

BIOPHYSICAL MEASUREMENTS

BY
PETER STRONG

Illustrations
by
DOROTHY FREED



MEASUREMENT CONCEPTS

FIRST EDITION, THIRD PRINTING JUNE 1973

062-1247-00

© 1970, TEKTRONIX, INC.
BEAVERTON, OREGON 97005
ALL RIGHTS RESERVED

CONTENTS

CAUTION

SCOPE 1

SECTION I	PHYSIOLOGY AND GENERATION OF BIOMEDICAL POTENTIALS WITHIN MAN	5
1	THE CELL AS A BIOELECTRIC GENERATOR	7
	1.1 The Source of Internal Cell Potentials	7
	1.2 Cell Stimulation and Stimulus Threshold	16
	1.3 Current from a Single Cell and the Resultant Externally Recorded Action Potential	16
	1.4 Externally Recorded Action Potential from a Group of Cells, the Travelling Wave of Depolarization	18
2	THE HEART AND THE CIRCULATORY SYSTEM	23
	2.1 The Cardiovascular Circulatory System	23
	2.2 The Heart	26
	2.3 Electrical Potentials Generated Within the Heart – Generation of the Electrocardiogram Waveform (ECG)	26
3	MUSCLE ACTION – AND THE SENSORY SYSTEM	31
	3.1 The Motor Unit	31
	3.2 Muscle Action	32
	3.3 The Muscular Servo-mechanism	33
	3.4 Reflex Response	34
	3.5 The Potential Generated During Muscle Action	34
	3.6 The Sense Receptors	36
	3.7 The Potential Generated by Sense Receptor Stimulation	36
4	THE BRAIN AND THE CENTRAL NERVOUS SYSTEM	39
	4.1 Nerve Cells in the Central Nervous System and in the Brain	39
	4.2 The Brain	40
	4.3 Excitation and Inhibition Potentials	42
	4.4 Evoked Potentials	45
	4.5 The Electroencephalogram (EEG)	45

SECTION II MEASUREMENT TECHNIQUES 47

- 5 ELECTROCARDIOGRAPHY (ECG) 49
 - 5.1 Electrocardiographic Planes 49
 - 5.2 Frontal Plane ECG Measurements 50
 - 5.3 Bipolar Limb Lead Frontal-plane ECG Measurements 52
 - 5.4 Unipolar Limb Lead Frontal-plane ECG Measurements 54
 - 5.5 Frontal-plane Axis Derivation 57
 - 5.6 Frontal-plane Electrode Positioning 57
 - 5.7 Transverse-plane ECG Measurements 60
 - 5.8 Sagittal plane ECG Measurements 63
 - 5.9 ECG Instrumentation Requirements 63
 - 5.10 Interpretation of the Electrocardiogram 65

- 6 VECTORCARDIOGRAPHY (VCG) 67
 - 6.1 The Spatial Vectorcardiogram 68
 - 6.2 Electrode Placement 69
 - 6.3 Frank Electrode System 70
 - 6.4 Polarity Convention 73
 - 6.5 Other Electrode Systems 74
 - 6.6 The Normal Vectorcardiogram 74
 - 6.7 Timing and Direction Reference 77
 - 6.8 Instrumentation Requirements 77

- 7 FETAL ELECTROCARDIOGRAPHY 81
 - 7.1 The Existence of the Fetal ECG 81
 - 7.2 The Normal Fetal ECG 82
 - 7.3 Subject Preparation 82
 - 7.4 Electrode Placement 83
 - 7.5 Electrical Interference 85
 - 7.6 Recording Techniques 86
 - 7.7 Interpretation of the Fetal ECG 89

- 8 BLOOD PRESSURE AND FLOW 93
 - 8.1 Direct Blood Pressure Measurement 93
 - 8.2 Indirect Blood Pressure Measurement 101
 - 8.3 Indirect Relative Blood Pressure Measurement 104
 - 8.4 Blood Flow Measurement 108
 - 8.5 Cardiac Output 112
 - 8.6 Blood Volume 113

- 9 RESPIRATION AND TEMPERATURE 115
 - 9.1 Physiological Considerations 115
 - 9.2 Respiratory Activity 118
 - 9.3 Respiratory Air Flow 122
 - 9.4 Respiratory Volume 125
 - 9.5 Temperature 129

10	ELECTROENCEPHALOGRAPHY	131
	10.1	The Characteristics of the Normal EEG 131
	10.2	EEG Electrode Considerations 134
	10.3	EEG Recording Instruments 136
	10.4	EEG Recording Modes 139
	10.5	Unusual EEG Display Modes 140
	10.6	Intra-cranial Electrode Placement 143
	10.7	Applications of the EEG 144
	10.8	The Characteristics of the Abnormal EEG 145
	10.9	Intentional Modification of a Subject's EEG 147
11	EVOKED CORTICAL RESPONSES	149
	11.1	Evoked Action Potential 149
	11.2	Microelectrode Technique 150
	11.3	Input Neutralized Amplification 157
	11.4	Noise Reduction – Averaging 159
	11.5	Typical Evoked Responses 164
	11.6	Stimulation 166
	11.7	Sterotaxic Instruments 166
	11.8	The Electroretinogram – ERG 169
12	STIMULATION – ELECTROMYOGRAPHY – NERVE CONDUCTION	171
	12.1	Stimulation 171
	12.2	Stimulus Isolation 175
	12.3	Strength/Duration Curves 176
	12.4	Myography 180
	12.5	Electromyography – EMG 182
	12.6	Electromyography with Voluntary Muscular Action 182
	12.7	Electromyography during Electrical Stimulation 185
	12.8	The H Reflex 187
	12.9	Nerve Conduction 189
	12.10	Repetitive Stimulation 192
	12.11	Smooth Muscle Potentials 193
13	GALVANIC SKIN REFLEX – GSR	195
	13.1	The Autonomic Nervous System 195
	13.2	GSR Measurement by Resistance Change 197
	13.3	GSR Measurement by Potential Detection 198
	13.4	Electrical Skin Resistance 199
14	ULTRASONOGRAPHY	201
	14.1	Ultrasonic Systems 203
	14.2	“A” Scan Ultrasonography 204
	14.3	“Time-motion” Mode Ultrasonography 207
	14.4	Ultrasonic Scanning 209
	14.5	Doppler Ultrasound 211

SECTION III	INSTRUMENTATION	213
15	INSTRUMENTATION SUMMARY	215
	15.1 Biomedical Signals	216
16	ELECTRODES	219
	16.1 Electrode Offset Potential	219
	16.2 Electrode Offset Potential Characteristics	225
	16.3 Other Electrode Characteristics	232
	16.4 Reusable Surface Electrodes	233
	16.5 Disposable Surface Electrodes	240
	16.6 Needle Electrodes	240
	16.7 Microelectrodes	243
	16.8 Electrode Application	245
	16.9 Stimulating Electrodes	247
	16.10 Comparison of Electrode Types	247
17	GROUNDING — SAFETY	249
	17.1 Grounding	249
	17.2 Induced Ground Currents	252
	17.3 Electrostatically Induced Currents	255
	17.4 Electric Shock Current Thresholds	255
	17.5 Instrumentation Safety Considerations	259
	17.6 Electrical Service Grounding	262
	17.7 Operating Room Isolation	264
	17.8 Electrocautery and Defibrillation	265
18	TRANSDUCERS — TRANSDUCER SYSTEMS	269
	18.1 Resistive Transducer Concepts	270
	18.2 The Unbalanced Wheatstone Bridge	273
	18.3 Practical Transducer Systems using the Unbalanced Wheatstone Bridge	276
	18.4 AC- and DC-bridge Systems	278
	18.5 Displacement Transducers	283
	18.6 Force, Pressure and Acceleration Transducers	288
	18.7 Transducers for Nonmechanical Quantities	291
19	AMPLIFIERS	293
	19.1 The Differential Amplifier used with Electrodes	295
	19.2 Common Mode Rejection	295
	19.3 Input Resistance	300
	19.4 Input Guarding	302
	19.5 Input Current	306
	19.6 Dynamic Range, DC Offset and Recovery	308
	19.7 Noise and Drift	310
	19.8 Specialized Amplifiers	317

20	SIGNAL PROCESSORS — OPERATIONAL AMPLIFIERS	319
20.1	Optimum Bandwidth for Physiological Signals	319
20.2	Amplifier Noise Reduction by Bandwidth Limiting	321
20.3	Amplifier Low Frequency Response Limiting	325
20.4	Line Frequency Rejection	325
20.5	Noise Reduction by Signal Averaging	326
20.6	Operational Amplifiers	326
20.7	Operational Amplifier Applications	330
20.8	Integration and Differentiation with Operational Amplifiers	334
21	OSCILLOSCOPES	337
21.1	Oscilloscope Vertical Amplifiers	338
21.2	Oscilloscope Horizontal Amplifiers/Sweep Generators	339
21.3	Oscilloscope CRT Displays	341
21.4	Typical Oscilloscopes	342
21.5	Rate Intensification	349
21.6	Slave Oscilloscopes	351
22	PULSE GENERATORS AND STIMULATORS	361
22.1	Stimulus Systems	363
22.2	Tektronix 160 Series Pulse Generators	364
22.3	High <i>E</i> or <i>I</i> Output via a Power Operational Amplifier	367
22.4	High <i>E</i> or <i>I</i> Output via a Stimulus Isolation Unit	369
22.5	Cardiac Pacemakers	370
22.6	Cardiac Defibrillators	373
23	DISPLAY DEVICES AND INDICATORS	375
23.1	Tektronix Display Units	377
23.2	Resolution	383
23.3	Large Screen Expanded-sweep Displays	385
23.4	Indicators	387
24	OSCILLOSCOPE CAMERAS	389
24.1	Conventional CRT Photography	391
24.2	Photography of Curved Faceplate CRT's	393
24.3	Continuous Motion Cameras	395
25	GRAPHIC RECORDERS	399
25.1	Basic Recorder Mechanisms	403
25.2	Recording Formats	405
25.3	Writing Principles	407
25.4	Commercial Galvanometric Recorders	409
25.5	Special Purpose Recorders	410

- 26 MAGNETIC TAPE RECORDERS 413
 - 26.1 Direct Magnetic Tape Recording 415
 - 26.2 Indirect FM Magnetic Tape Recording 417
 - 26.3 Tape Transport Mechanism 419
 - 26.4 Other Magnetic Tape Recorder Considerations 420

- 27 DATA TRANSMISSION AND PROCESSING 423
 - 27.1 Data Transmission via Shielded Cable 423
 - 27.2 Data Transmission via a Telemetry Link 426
 - 27.3 Time Division Multiplexing 429
 - 27.4 Data Processing with an Analog Computer 430
 - 27.5 Data Processing with a Digital Computer 432
 - 27.6 Data Processing Applications and Software 435

- 28 INTENSIVE CARE CONCEPTS 437
 - 28.1 Physiological Functions to be Monitored during Intensive Care 438
 - 28.2 Intensive Care Instrumentation 440

SECTION IV APPENDIX 447

- 29 CUSTOM INSTRUMENTATION 449
 - 29.1 Type 410 Modification for Fetal ECG Use 450
 - 29.2 A BNC Input Adapter for the Type 410 Monitor 453
 - 29.3 A Signal Output Adapter for the Type 410 Monitor 453
 - 29.4 A Resistive Transducer Adapter for the Type 410 Monitor 454
 - 29.5 A Thermistor Pneumograph for the Type 410 Monitor 457
 - 29.6 An Input Neutralizing Adapter for the Type 3A8 Operational Amplifier 458
 - 29.7 An Absolute Value Adapter for the Type 3A8 Operational Amplifier 460
 - 29.8 A Low Speed Gating Adapter for the Type 3A8 Operational Amplifier 463
 - 29.9 A Resetting Step Generator Adapter for the Type 3A8 Operational Amplifier 465
 - 29.10 A Self-contained Resetting Stairstep Generator 466
 - 29.11 A Frank Network for Vectorcardiographic Use 469
 - 29.12 A Current Limiting Adapter for Protection from Electric Shock 470
 - 29.13 A Lo-pass Filter for Physiological Signal Processing 471
 - 29.14 A Pulse Shaping Circuit Simulating the Action Potential 473
 - 29.15 A Constant Current Pulse Source for GSR and Other Uses 474

- 30 DEFINITIONS 475

ACKNOWLEDGMENTS

Professor Harold W. Shipton
Bioengineering Resource Facility
University of Iowa
Iowa City, Iowa, U.S.A.

Dr. David J. Dewhurst
Department of Physiology
University of Melbourne
Melbourne, Vic., Australia

The author gratefully acknowledges the assistance provided by Dr. Dewhurst and Professor Shipton in the writing of this book. Professor Shipton's contributions are particularly evident in chapters four and ten dealing with the electroencephalogram. Dr. Dewhurst's extensive contributions, specially to the first four chapters on physiology, were invaluable.

CAUTION

Engineers and nonmedically qualified personnel should not perform biophysical measurements on human subjects unless such measurements are conducted under professional supervision. Many of the measurement techniques described in this book, while incorporating medically accepted procedures, are not entirely free from risk. These risks may be minimized by following the precautions outlined in Chapter 17, "Grounding - Safety."

One or more "current limiting adapters for protection from electric shock" as described in Chapter 29, Section 29.12, should be used between human subjects and electronic instrumentation to protect the subject from electric shock should a failure occur within this instrumentation.

SCOPE

It is the intent of this publication to familiarize engineering personnel with electronic measurements associated with the biophysical sciences. As such, a developed knowledge of electronics and electronic measuring techniques is assumed, but biological and physiological aspects are presented from first principles and simplified to a level consistent with the understanding of the basic principles involved.

This book will be found to differ somewhat from other publications on "medical electronics." Many of these publications are authored by, and directed toward, medical personnel. As such, engineering personnel may find the electronics aspects somewhat oversimplified and the physiological aspects taken for granted. The opposite approach has been taken here. The following books are suggested as general references on Biophysical Measurements. Specific references are given, where appropriate, in the text throughout this book.

Dewhurst, *Physical Instrumentation In Medicine and Biology*, London: Pergamon, 1966.

Dickinson, *Electrophysiological Technique*, London: Electronic Engineering, 1950.

Donaldson, *Electronic Apparatus For Biological Research*, London: Butterworth, 1958.

Geddes and Baker, *Principles of Applied Biomedical Instrumentation*, New York: Wiley & Sons, 1968.

Kay, *Experimental Biology*, London: Chapman and Hall, 1964.

Stacey, *Biological and Medical Electronics*, New York: McGraw-Hill, 1960.

Suckling, *Bioelectricity*, New York: McGraw-Hill, 1961.

Yanof, *Biomedical Electronics*, Philadelphia: Davis, 1965.

Engineers interfacing with the medical profession are encouraged to learn as much as possible about medical and hospital practice and in particular about the physiology of the human body. It is only by gaining such an understanding that they can communicate intelligently with members of the medical profession. One is often approached by a doctor with the request, "can you supply me with this item of equipment?" The tactful but firm reply must be, "what do you want it to do?" Very often the requirement, once understood, can be met far more effectively by some more modern development than that embodied in the original request. In this way a real service to the medical profession is provided and the engineer becomes a colleague rather than a graduate technical assistant.

Information is separated into the following categories:

SECTION I -- Physiology and generation of bioelectric potentials within man.

SECTION II -- Measurement techniques required to perform various biophysical measurements

SECTION III -- Instrumentation required to implement the measurement techniques covered in Section II.

A reader wishing to study, for example, electrocardiographic techniques should thus first study Section I, Chapter 2, to determine the source of the bioelectric potential referred to as the electrocardiogram; he should then study Section II, Chapter 5, to determine the measurement techniques necessary in recording this electrocardiogram and, finally, he should study Section III to determine the instrumentation necessary to implement these measurement techniques.

Wherever possible, the biophysical measurements covered assume man as the subject. Although, in practice, many of these measurements are performed on laboratory animals, such as rats, cats, dogs, and monkeys, the principles involved and the results obtained relate directly to man. No division is attempted between biophysical measurements performed in a clinical environment and biophysical measurements performed in a research environment as they are, from the electronic engineering viewpoint, essentially similar. In general, only commonly accepted and widely used measurement techniques are covered in this publication as it would be impossible to document the infinite variety of unique, specialized, biophysical measurements being performed by different clinicians and researchers in the biophysical sciences. This publication should not be interpreted as a general text on medical electronics; it is, as its title implies, limited to measurement techniques.

SECTION I

PHYSIOLOGY AND GENERATION OF BIOELECTRIC POTENTIALS WITHIN MAN

The following four chapters (1-4) cover the basic physiology of the cell, the heart, the muscular system and the brain and the generation of electrical activity within these physiological systems.

The intent of these four chapters is to familiarize engineering personnel with the physiological aspects associated with the various bioelectric generators within the body. In general, the information is presented in the simplest possible form. Thus, while satisfying the intent of this publication, the professional physiologist may find the subject material somewhat oversimplified.

Interested readers will find the following texts invaluable for further study:

Brazier, *The Electrical Activity of the Nervous System*, Baltimore: Williams and Wilkins, 1968.

Eccles, *The Neurophysiological Basis of Mind*, London: Oxford, 1953.

Eccles, *The Physiology of Nerve Cells*, New York: John Hopkins, 1957.

Hoffman and Cranefield, *Electrophysiology of the Heart*, New York: McGraw-Hill, 1960.

Katz, *Nerve, Muscle and Synapse*, New York: McGraw-Hill, 1966.

1

THE CELL AS A BIOELECTRIC GENERATOR

The cell is the basic source of all bioelectric potentials. A bioelectric potential may be defined as the difference in potential between the inside and the outside of a cell; in other words, the difference in potential existing across the cell wall or membrane. A cell consists of an ionic conductor separated from the outside environment by a semipermeable or selectively permeable cell membrane. Many different types of cells comprise any one species of living matter. Human cells may vary from 1 micron to 100 microns in diameter, from 1 millimeter to 1 meter in length and have a typical membrane thickness of 100 Angstrom units. One micron is equal to 10^{-4} centimeters and one Angstrom unit is equal to 10^{-8} centimeters. Bioelectricity is studied both from the viewpoint of the source of electrical energy within the cell and also from the viewpoint of the laws of electrolytic current flow relative to the remote ionic fields produced by the cell. In electrophysiology we may, in some cases, penetrate a cell to investigate its internal potential. More commonly, however, we make measurements external to a group of cells while these cells are supplying electrolytic current flow.

1.1 THE SOURCE OF INTERNAL CELL POTENTIALS

Experimental investigations with microelectrodes have shown that the internal resting potential within a cell is approximately -90 millivolts with reference to the outside of the cell. This potential changes to approximately +20 millivolts for a short period during cell activity. Cell activity results from some form of stimulation as described later in this chapter. The Hodgkin-Huxley theory, initially postulated during the 1950's, is generally considered to give the best explanation as to the source of these potentials and provides equations that give an empirical mathematical fit to experimental data. This theory is briefly described as follows.

chemical
gradient

The interior of a cell primarily contains concentrations of sodium and potassium ions. These concentrations within a cell differ markedly from the concentrations of these ions in the space outside the cells (see Fig. 1-1). Elementary ionic theory states that, under suitable conditions, any uneven distribution of ionic concentration in an aqueous solution will result in a potential difference between the regions of different concentration. If, for example, solutions containing unequal concentrations of ions are separated by a membrane semipermeable to these ions, a potential will be found to exist (see Fig. 1-2). This potential, which is referred to later as the chemical gradient, is given by the Nernst relation:

Nernst
relation

Potential (mV) =

$$61.6 \text{ Log } \frac{\text{Concentration, one side of membrane}}{\text{Concentration, other side of membrane}}$$

Or, for uni-univalent ionic solutions, the Nernst relation simplifies to:

$$\text{Potential (mV)} = 61.6 \frac{U - V}{U + V}$$

where U = Mobility of the negative ions
(anions) through the membrane

V = Mobility of the positive ions
(cations) through the membrane

Referring to Fig. 1-2, for a 10:1 activity (concentration) ratio at 37°C, the relative mobilities of the chloride and sodium ions are 65.4 and 43.6 respectively. Applying these values to the Nernst relation gives:

$$\begin{aligned} \text{Potential (mV)} &= 61.6 \frac{65.4 - 43.6}{65.4 + 43.6} \\ &= 12 \text{ mV} \end{aligned}$$

This can be confirmed with a voltmeter as shown in Fig. 1-2. This potential will, of course, run down as diffusion proceeds, unlike that of a living cell.

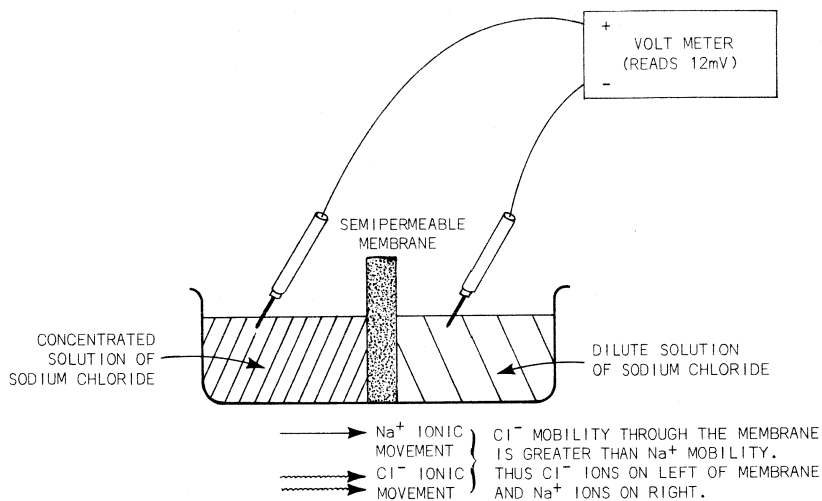


Fig. 1-1. Potential generated by ionic concentration difference between two solutions.

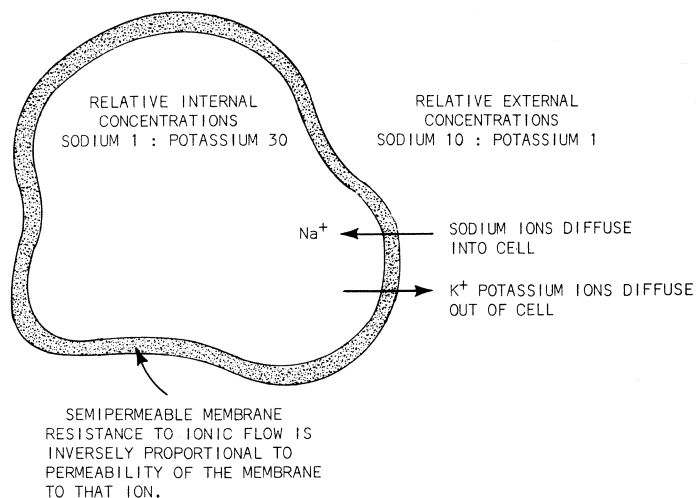
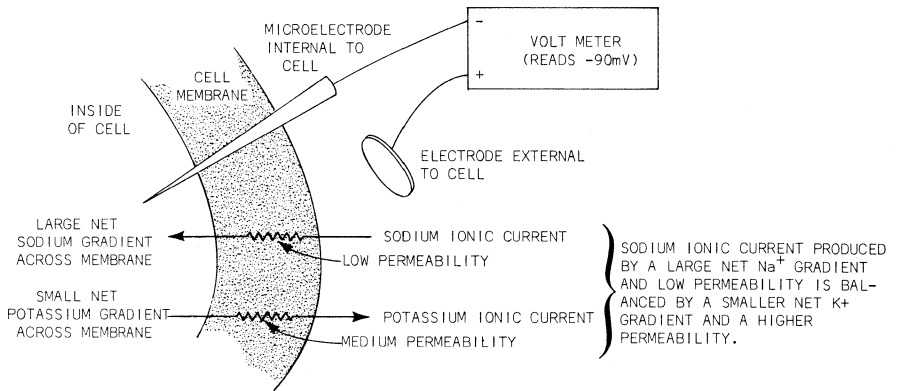
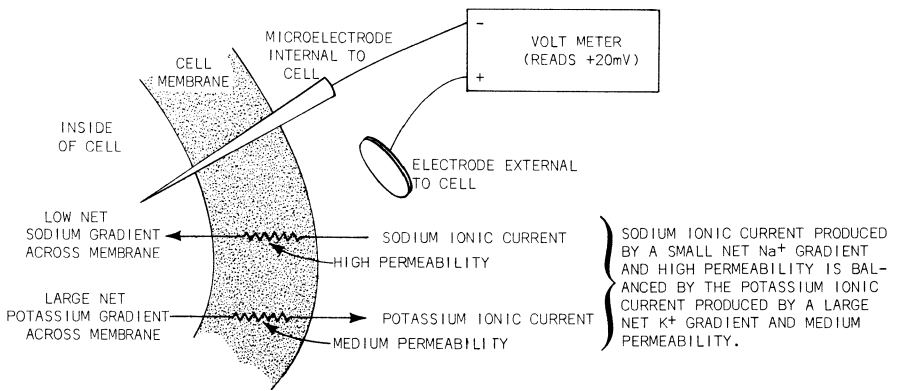


Fig. 1-2. Typical concentrations of sodium and potassium ions within a cell.



(A) IONIC CURRENTS INVOLVED IN A POLARIZED CELL



(B) IONIC CURRENTS INVOLVED IN A DEPOLARIZED CELL

Fig. 1-3. Cell ionic currents.

net
gradient

The ionic current produced by ion movement through a semipermeable membrane depends on the permeability of the membrane and also on the "gradient" that forces the ion through the membrane. This gradient, referred to as the net gradient, consists of both a chemical gradient and an electrical gradient. A chemical gradient is formed due to a difference in concentration producing a potential gradient as given by the Nernst relation. An electrical gradient is formed as a result of a potential that may exist across the membrane due to some other source.

cell
concentra-
tions

Experimental investigations have shown that a marked difference in concentrations of both sodium and potassium ions exists across a cell membrane. In mammalian nerve cells as shown in Fig. 1-1, the concentration of potassium ions is in the vicinity of 30 times higher inside the cell than in the fluid external to the cell. On the other hand, sodium ions are approximately 10 times more concentrated in the fluid external to the cell than in the fluid within the cell.

cell in
resting
state

Consider a cell in its resting or polarized state (Fig. 1-3A). In this state the membrane is moderately permeable to potassium ions, that is, potassium ions can pass fairly readily through the membrane as the membrane offers medium resistance. This membrane is, however, almost impermeable to sodium ions and, thus, offers a high resistance to the passage of these ions. A large net gradient affects the movement of sodium ions into the cell. This net gradient consists of a chemical gradient produced by the 10-to-1 concentration difference between sodium ions on each side of the membrane and a 90 mV electrical gradient produced by the standing potential within the cell. The net gradient affecting the movement of potassium ions out of the cell is considerably less than the net sodium gradient. This gradient consists of a large chemical gradient due to the 30-to-1 concentration difference across the membrane; however, this chemical gradient is opposed by the electrical gradient produced by the 90 mV standing potential within the cell. Thus, although the membrane is almost impermeable to sodium ions, the net sodium gradient is high. Conversely, although the membrane is moderately permeable to potassium ions, the net potassium gradient is low. The net result is that

net
gradient

-90 mV
resting
level
polarized
cell

the sodium and potassium currents are equal; the sodium current balances the potassium current with a resultant current of zero. Since the net current through the membrane is zero, the cell's internal potential will not change and will remain at its -90 mV resting level. Indeed, this -90 mV resting level is determined by the internal cell potential required for sodium and potassium current balance.

cell after
stimulus

depolarization

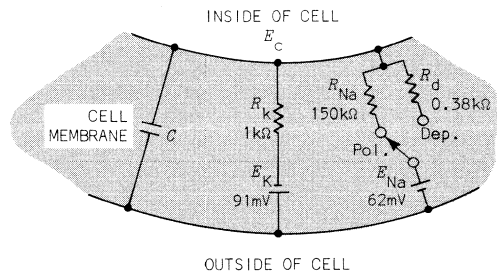
When the cell receives a stimulus from an outside source, the characteristics of the membrane at the point of stimulation will be markedly altered and, thus, the ionic currents will also change. After stimulation, the membrane permeability to potassium ions is unaltered but the permeability to sodium ions is increased. A much lower resistance is offered to the flow of sodium ions, thus increasing the sodium ionic current. This increased sodium ionic current causes more positive ions to pass into the cell than are passing out of the cell, causing the internal cell potential to drop from -90 mV in an attempt to achieve sodium current and potassium current balance. As this potential decreases, the net sodium gradient across the membrane decreases and the net potassium gradient across the membrane increases, causing the currents to decrease and increase, respectively. This process continues until current balance is again obtained, at which time the internal cell potential is +20 mV. The cell is then referred to as being in a depolarized state.

repolarization

By the time the cell has fully depolarized the characteristics of the membrane have begun to revert back to their prestimulus state. This causes the sodium ionic current to be considerably lower than the potassium ionic current; the internal cell potential thus begins to go negative with the process continuing until the -90 mV resting potential of the cell is once again obtained.

cell
electrical
analogy

An electrical analogy to a cell membrane is shown in Fig. 1-4. This circuit cannot strictly be referred to as an equivalent circuit as the electronic current flow in an electrical circuit and the ionic current flow through a cell membrane cannot be said to be equivalent. After assigning resistance values inversely proportional to the relative permeability of the membrane and assuming potassium and sodium concentration ratios, then the intracellular potential for both a polarized cell and a depolarized cell can be determined. The values assumed are analogous to actual values found in a cell.



E_C = INTRACELLULAR POTENTIAL WITH RESPECT TO THE OUTSIDE OF THE CELL.

E_K = NERNST POTENTIAL DUE TO THE POTASSIUM ION CONCENTRATION DIFFERENTIAL ACROSS THE MEMBRANE

E_{Na} = NERNST POTENTIAL DUE TO THE SODIUM ION CONCENTRATION DIFFERENTIAL ACROSS THE MEMBRANE.

R_K = RELATIVE PERMEABILITY OF THE MEMBRANE TO THE FLOW OF POTASSIUM IONS THROUGH IT.

R_{Na} = RELATIVE PERMEABILITY OF THE MEMBRANE TO FLOW OF SODIUM IONS THROUGH IT WHEN THE CELL IS POLARIZED.

R_d = RELATIVE PERMEABILITY OF THE MEMBRANE TO THE FLOW OF SODIUM IONS THROUGH IT WHEN THE CELL IS DEPOLARIZING.

C = CAPACITY OF THE CELL.

Fig. 1-4. An electrical analogy to a cell membrane.

Assume:

Relative values for R_K , R_{Na} and R_D are $1\text{ k}\Omega$, $150\text{ k}\Omega$ and $0.35\text{ k}\Omega$ respectively.

Potassium ion concentration ratio of 30:1 inside to outside.

Sodium ion concentration ratio of 10:1 outside to inside.

Then:

$$E_K = 61.6 \text{ Log } \frac{30}{1} = 91 \text{ mV} \quad \text{-- by Nernst relation.}$$

$$E_{Na} = 61.6 \text{ Log } \frac{10}{1} = 62 \text{ mV in opposite polarity to } E_K \text{ -- Nernst.}$$

For a polarized cell:

$$\text{net potassium current} + \text{net sodium current} = 0$$

$$\frac{\text{net potassium gradient}}{R_K} + \frac{\text{net sodium gradient}}{R_{Na}} = 0$$

$$\frac{\text{potassium chemical gradient} + \text{potassium electrical gradient}}{R_K} + \frac{\text{sodium chemical gradient} + \text{sodium electrical gradient}}{R_{Na}} = 0$$

$$\frac{E_K}{R_K} + \frac{E_c}{R_K} + \frac{-E_{Na} + E_c}{R_{Na}} = 0$$

$$\frac{91 \times 10^{-3} + E_c}{1 \times 10^3} + \frac{-62 \times 10^{-3} + E_c}{150 \times 10^3} = 0$$

Solving:

$$E_c = 90 \times 10^{-3} = -90 \text{ mV (polarized)}$$

Similarly, for a depolarized cell, R_{Na} is replaced by R_D .

$$E_c = +20 \text{ mV (depolarized)}$$

cell action
potential

If a microelectrode were inserted into the cell as shown in Figs. 1-3A and 1-3B and a stimulus were applied to the cell, the output of the microelectrode would appear as shown in Fig. 1-5. This waveform is known as the "cell action potential." It should be noted that the currents involved in bioelectricity are unlike the currents involved in electronics. Bioelectric currents are

due to positive and negative ion movement within a conductive fluid. As these ions possess finite mass and encounter resistance to movement within the fluid their speeds are limited. The cell action potential, thus, shows a finite "risetime" and "falltime."

sodium-
potassium
pump

The ionic concentration gradient across the cell membrane is maintained by virtue of metabolic energy expended by the cell in "pumping" ions against the ionic gradient formed by the differing ionic concentrations between the inside and outside of the cell. This action has been referred to as the "sodium-potassium pump."

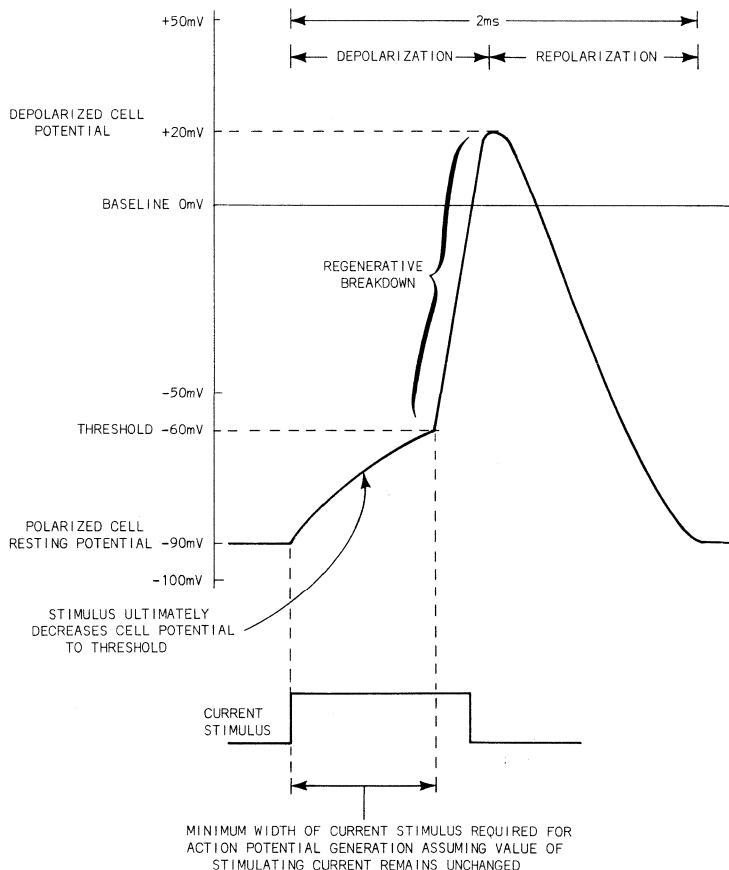


Fig. 1-5. Cell action potential (internally recorded with microelectrode).

1.2 CELL STIMULATION AND STIMULUS THRESHOLD

stimulus
threshold

A cell may be stimulated, or caused to depolarize and then repolarize, by subjecting the cell membrane to an ionic current. This current may be produced by other cells, it may be produced by ionic currents existing as nerve impulses, or it may be artificially produced by some external current stimulus. A cell will be stimulated when sufficient positive ions are added to the inside of the cell to cause its resting potential to be decreased from its -90 mV level to approximately -60 mV. Once this threshold level is reached, the cell depolarizes without requiring the addition of any further positive ions to the inside of the cell from the stimulus source. Unless a stimulus above a certain minimum value is received, known as the stimulus threshold, the cell will not be depolarized and no action potential will be generated. The stimulus required to exceed the threshold is a function of both current and time; the threshold may be exceeded by a short, high-current pulse or by a longer, lower-current pulse (see Fig. 1-5).

refractory
period

Since the energy associated with the action potential is developed from metabolic processes within the cell itself and not from the stimulus, a finite period of time, known as the refractory period, is required for metabolic processes within the cell to return the cell to its prestimulus state. This refractory period has been observed in most cells found in the nervous system. Closer study reveals that the refractory period has two parts: The first in which no stimulus, however strong, will cause depolarization (the absolute refractory period) and the second when depolarization occurs only if the stimulus is of more than normal threshold strength (the relative refractory period).

1.3 CURRENT FROM A SINGLE CELL AND THE RESULTANT EXTERNALLY RECORDED ACTION POTENTIAL

We have previously discussed the depolarizing and repolarizing action of cells and the resulting potential existing within the cell. As stated earlier, this potential may be recorded with a microelectrode; however, in most bioelectric

external
electrodes

measurements, this potential is recorded by electrodes external to the actual cell. These external electrodes typically would record the net action of many hundreds of cells, but for the time being, we will consider only a single cell. When recording with external electrodes, an action potential is produced between these electrodes during periods of current flow; that is to say, no potential exists when cells are either in their depolarized or repolarized state. A potential exists only while the cell is changing from one state to another. As the external action potential is generated by the external current that flows during cell activity, the shape of the action potential is related to the variation of this current with time.

external
action
potential

The external potential field rises to its maximum value sometime during the regenerative breakdown phase of the membrane. The external action potential that is recorded from the cell is somewhat similar to a mathematical time derivative of the transmembrane potential. This potential is detected with maximum amplitude when one electrode is placed as near as possible to the active area and the other electrode is located in a completely inactive or remote area. It is detected with reduced amplitude as the electrodes are placed closer to each other so they intercept smaller elements of potential difference.

depolarization
current

Consider a single polarized cell; the inside of the cell is negative with respect to the outside environment which may be regarded as a reference. As stated previously, the net ionic current flow across the membrane is zero; thus, ionic current flow to and from the cell is zero. Should these conditions be altered due to the presence of a stimulating current through the cell membrane, then regenerative membrane breakdown will occur and the cell will depolarize. During the depolarization process the net current through the membrane is not zero; there is a net positive ionic current into the cell through the cell membrane. This current may be detected as a potential difference between two electrodes placed in the vicinity of the cell with the potential difference being produced across the finite resistance of the fluids external to the cell.

This current continues until the cell is fully depolarized, thus, the potential will appear between the two electrodes while the cell is undergoing depolarization. For the short time that the cell exists in a depolarized state, the net current through the membrane is once again zero, thus, no potential will appear between the electrodes.

repolari-
zation
current

Almost immediately after depolarization the cell begins to repolarize again. During repolarization, positive ionic current flows from the cell membrane, that is, in the opposite direction to ionic current flow during depolarization. During this period of current flow, a potential will be detected between the electrodes and, since this current is in the opposite direction to the depolarization current, the potential produced between the electrodes during repolarization will be of the opposite polarity to the potential produced during depolarization. The area included by the depolarization and the repolarization potential waveforms is the same since the quantity of current involved in each process is the same. It should be noted that the process described above is somewhat theoretical as it describes the external action potential generated by a single cell. In the following discussion we show that single cell depolarization invariably results in depolarization of the adjacent cells and, hence, this externally recorded action potential will be the net summation of the results obtained from these cells.

1.4 EXTERNALLY RECORDED ACTION POTENTIAL FROM A GROUP OF CELLS, THE TRAVELLING WAVE OF DEPOLARIZATION

synchronous
depolari-
zation

The preceding discussion must be modified to allow for many cells in close proximity to one another and to allow for the appreciable length of many of these cells. Consider a group of cells in close proximity to one another as shown in Fig. 1-6 and 1-7. Under certain conditions of stimulation these cells may all depolarize at the same time (synchronous depolarization); however, the repolarization process is random. Repolarization of the individual cells will occur at different times. The resultant externally recorded action potential is shown in Fig. 1-6. Once again, the area included under each wave is the same since the quantity of current involved in each process is the same.

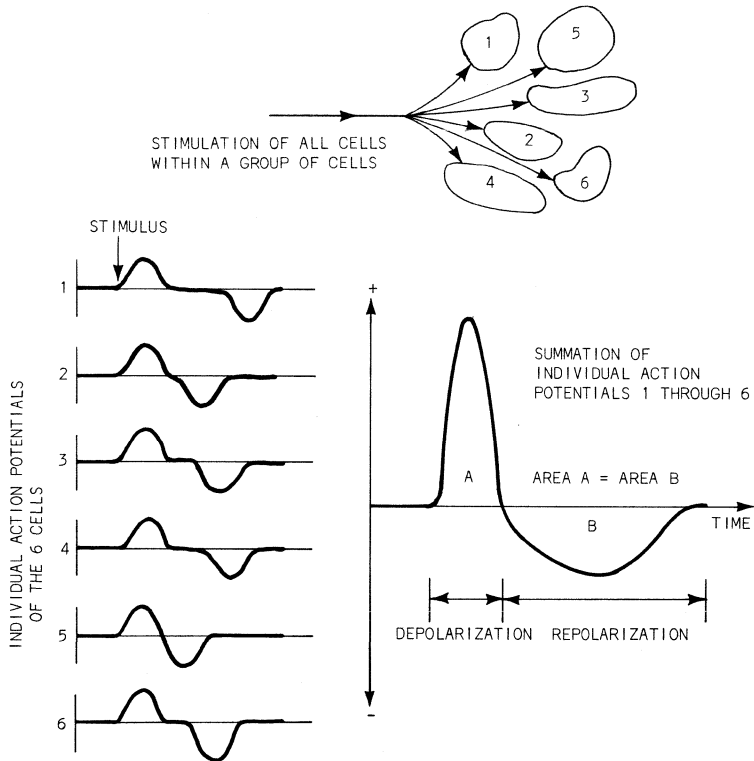


Fig. 1-6. External action potential produced by a group of cells all depolarizing as a result of one stimulation (synchronous depolarization).

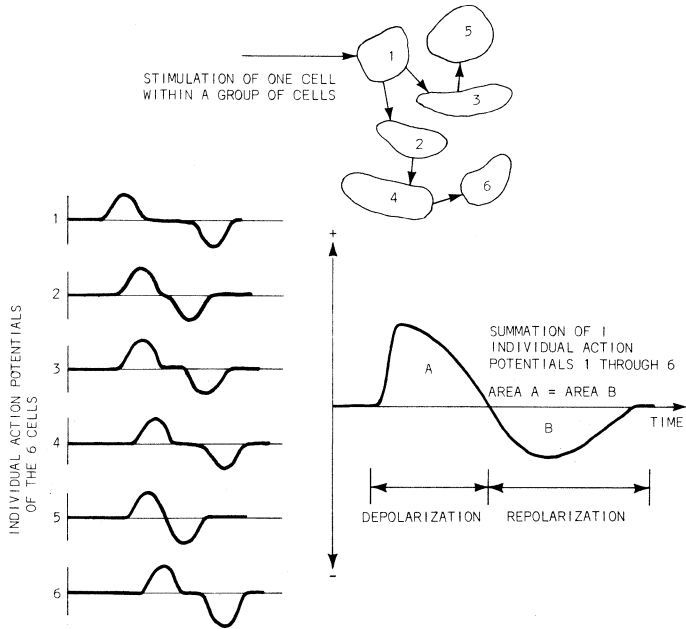


Fig. 1-7. External action potential produced by a group of cells depolarizing due to stimulation by adjacent cells (asynchronous depolarization).

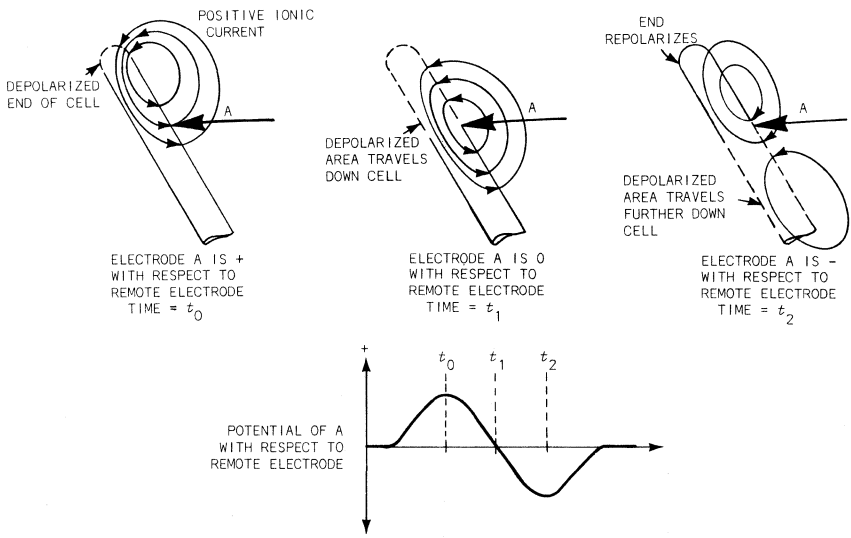


Fig. 1-8. "Traveling wave of depolarization" resulting from depolarization of successive section of a long cell.

asynchronous
depolarization

Under other conditions of stimulation, the group of cells described previously will not all depolarize at the same time (asynchronous depolarization). The stimulation may result in one cell depolarizing; the action of this cell depolarizing will then act as a stimulus on its adjacent cell causing it to depolarize also. This chain reaction would proceed until all cells in a particular area have depolarized. The resultant externally recorded action potential would appear as shown in Fig. 1-7. In practice, combinations of synchronous and asynchronous depolarization occur in a group of cells.

travelling
wave within
cells

In the same way as the depolarization of one cell causes adjacent cells to depolarize, depolarization of a localized area of an individual cell will cause depolarization of other parts of the same cell. Thus, the depolarization process will appear to travel along the length of the cell causing a travelling wave of depolarization as shown in Fig. 1-8.

As most bioelectric potentials are recorded as external cell action potentials, the results obtained are a summation of the action of many cells. The action potential waveform may be modified by the number of cells, the shape of these cells, and the type of stimulation applied.

2

THE HEART AND THE CIRCULATORY SYSTEM

myocardial
infarction

Coronary heart disease is the greatest single cause of death in most developed countries. In the United States alone, over a million persons die every year from heart disease; the most frequent cause of death is the acute myocardial infarction. Thus, a major percentage of medical effort, both in the research and clinical fields, is directed toward understanding of cardiovascular system and the subsequent prevention of heart disease.

2.1 THE CARDIOVASCULAR CIRCULATORY SYSTEM

cell
revitali-
zation

In a complex multicellular animal with a high density of cell population and a relatively small volume of tissue fluid bathing these cells, a continuous and rapid revitalization process must take place. This revitalization is the fundamental purpose of the cardiovascular circulatory system. The previous chapter discussed the potential produced within a single cell. This potential exists as a result of continuous metabolism within the cell. This metabolism process requires nutrients and excretes waste products; the circulatory system provides these nutrients and removes these waste products.

The major component of the circulatory system is the heart which supplies the power required to circulate blood throughout the body. The heart is two pumps in series; the smaller, right-hand section provides the power required to force blood through the lungs, and the larger, more powerful left-hand section provides the power required to force blood through the body. A simplified block diagram is shown in Fig. 2-1.

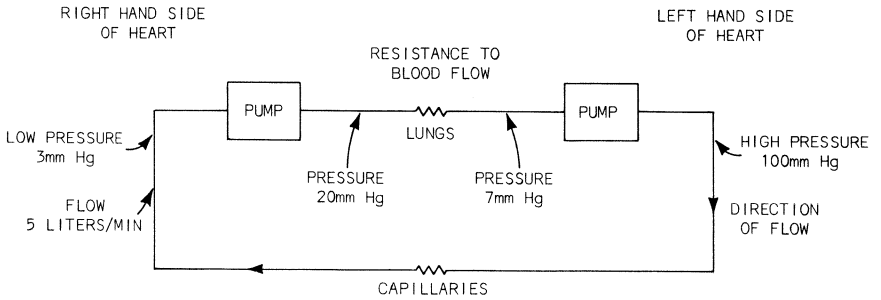


Fig. 2-1. Simplified block diagram of the circulatory system.

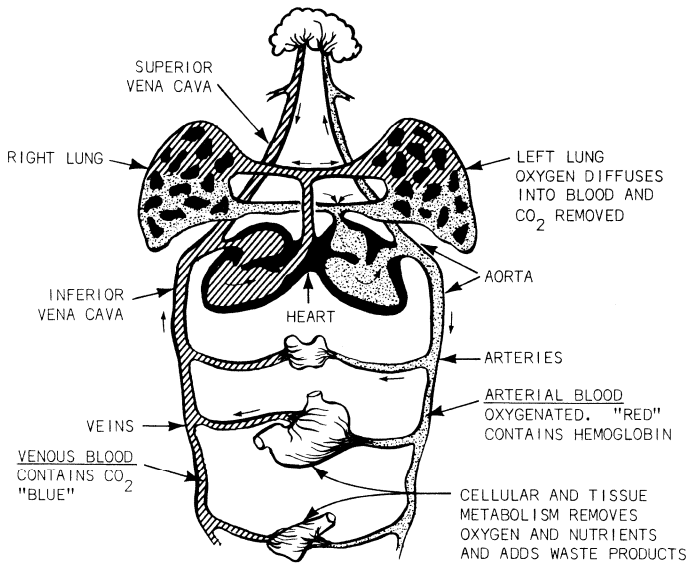


Fig. 2-2. The cardiovascular circulatory system.

blood flow

Elaborating on Fig. 2-1 and referring to Fig. 2-2, the blood flows from the heart into the aorta. The aorta curves in an arch up from the heart, down along the back bone and into the abdomen; from it other large arteries lead to the head, the digestive organs, the arms and the legs. From these arteries branch the smaller arterioles (or small arteries) and from these, branch billions of tiny capillaries. By the time the blood has reached the capillaries it is moving slowly along channels that are only about 10 microns in diameter. Here the blood discharges its load of dissolved food and oxygen to the body cells. These cells in turn deposit waste materials such as carbon dioxide into the blood stream. In yielding oxygen and taking on waste, the blood turns color from bright red to dull red or "blue." The blood now starts back to the heart passing from the capillaries into the venules. The venules converge into larger veins and then into the two largest veins just above and below the heart known as the vena cava. The blood empties into the right atrium, is pumped into the right ventricle and then moves out through the pulmonary artery to the lungs. The lungs supply the blood with fresh oxygen. The blood passes from the lungs to the left atrium, then is pumped into the left ventricle and passes out via the aorta to begin the circulation process again. This general flow throughout the body is known as the "systemic circulation;" the flow to and from the lungs is known as the "pulmonary circulation." Local circulations within the systemic system include the renal (to the kidneys), the hepatic portal (to the liver), the cerebral (to the brain) and the coronary (to the heart itself). The waste products contained in the blood are removed by the kidneys and liver (see Fig. 2-2). The average quantity of blood in man is about five liters and is completely circulated through the body in approximately one minute.

2.2 THE HEART

The heart itself weighs less than a pound, is about six inches long at its maximum dimension and lies pointed downward in the chest cavity to the left of the mid-center body line. The walls of the heart are made entirely of muscle; within these walls are four hollow chambers, a left and a right receiving chamber (atrium) and below them a left and a right pumping chamber (ventricle).

cardiac cycle	The cardiac cycle is characterized by the following mechanical events. Between beats, the heart mechanically rests and this is known as the period of diastole. During diastole the heart assumes its maximum size and fills with oxygenated blood returning from the lungs and venous blood returning from the body. The heart's period of mechanical activity is known as systole. The onset of systole is initiated by contraction of the muscles surrounding the atria which propels additional blood into the ventricles. The ventricles then begin to contract, thereby causing a rise in pressure within the ventricles. This increased pressure shuts the two atrioventricular valves and, with further contraction, the pressure continues to rise. Once the pressure of the systemic and pulmonary circulations are exceeded, a phase of ventricular ejection is begun. The aortic valve is forced open and blood is squeezed into the aorta and thence into the systemic circulation. Likewise, the pulmonary valve is forced open and blood is supplied to the pulmonary circulation. After the ventricular contents are partially ejected, the muscles surrounding the ventricles relax and the ventricular pressures fall. As soon as these pressures fall below the pressures sustained in the circulatory systems, the aortic and pulmonary valves close, signalling the onset of diastole.
diastole	
systole	

2.3 ELECTRICAL POTENTIALS GENERATED WITHIN THE HEART - GENERATION OF THE ELECTROCARDIOGRAM WAVEFORM (ECG)

The preceding section covers the mechanical activity of the heart and states that mechanical activity is initiated by contraction of the muscle surrounding the atria. The detailed relationship between cell, nerve and muscle producing this contraction is covered in Chapter 3. It is sufficient at this stage to state that muscle contraction is initiated by stimulation.

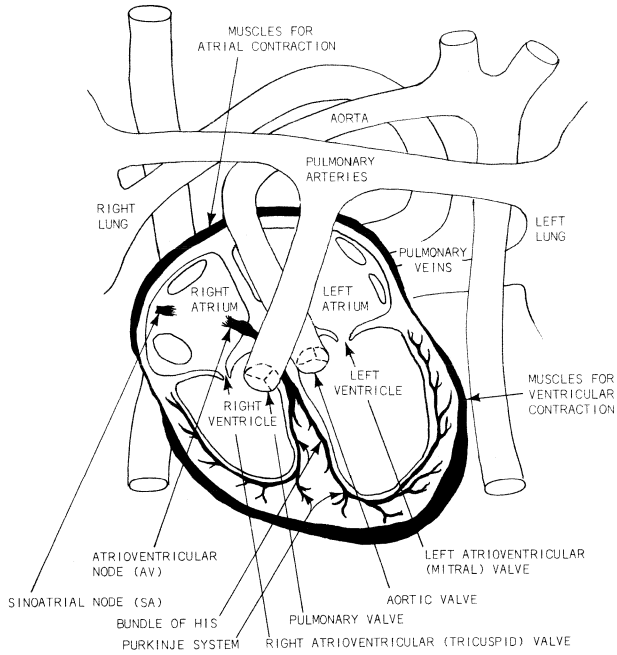


Fig. 2-3. The heart.

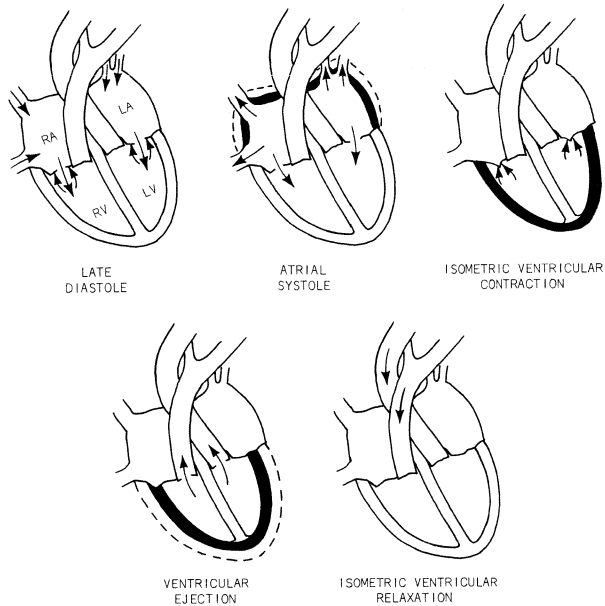


Fig. 2-4. Mechanical cycle of the heart.

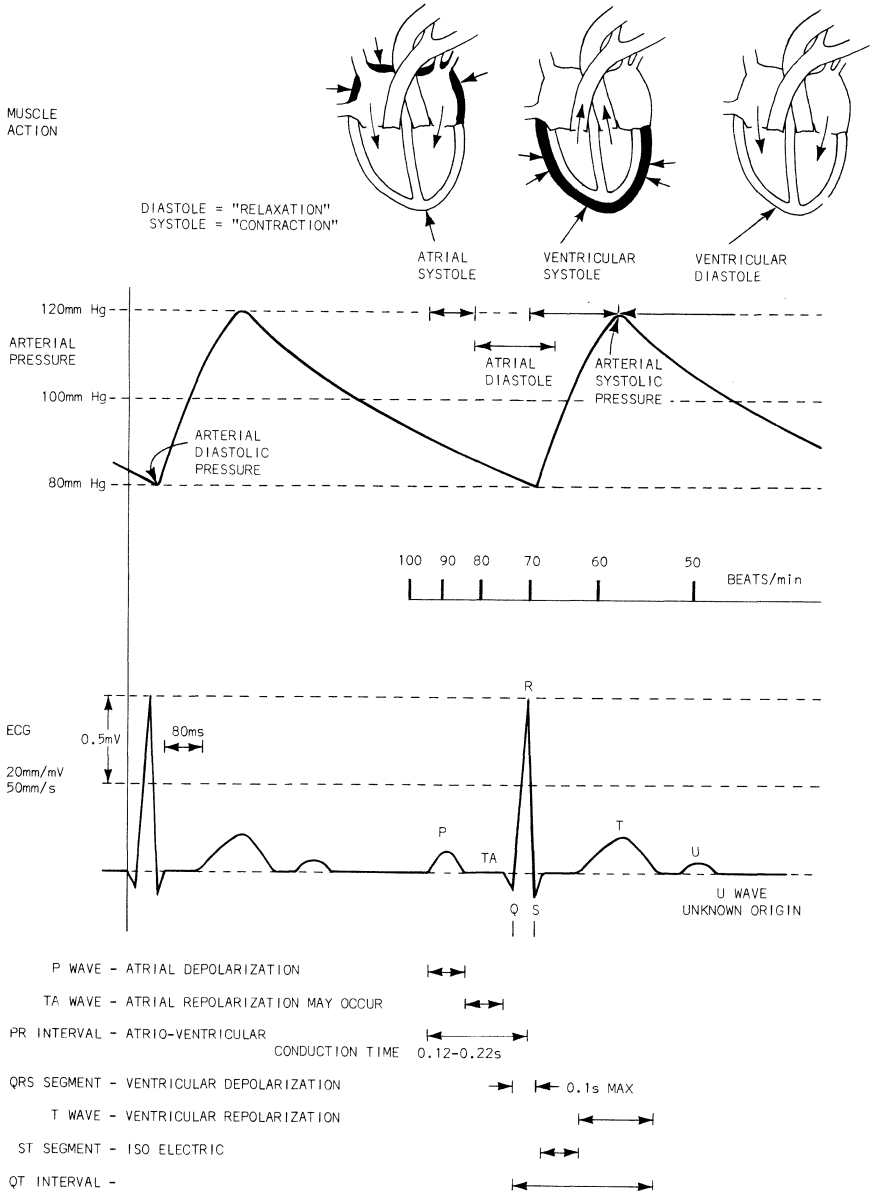


Fig. 2-5. The ECG waveform and related heart action.

sinoatrial
node

The right atrium contains a bundle of nerves known as the sinoatrial node (abbreviated SA node). This type of nerve cell is found nowhere else in the body. Its function is to start the heart beat and set its rhythm or pace. The electrical and mechanical output from the heart is initiated by stimulation from this node which results in contraction of the various heart muscles. Although the basic rhythm of the heart is self-sustaining through synchronization from the SA node within the heart, this rhythm is modified by certain nerve fibers external to the heart that affect the SA node. These nerves have a function in the normal control of the heart rate to respond to increased or decreased demand for blood by the body.

atrio-
ventricular
node

Impulses generated by the SA node stimulate contraction of the muscles comprising the atria. These impulses also travel along conducting fibers in the atrium to the atrioventricular node or AV node, stimulating depolarization of this node (Fig. 2-3). Stimulation of the atrioventricular node causes impulses to be sent to the myocardium or muscle comprising the ventricles via the bundle of His and the Purkinje conducting system resulting in contraction of this muscle. Thus, the muscular contractions necessary to maintain the heart's pumping action are initiated by depolarization and repolarization of the SA node and then depolarization and subsequent repolarization of the AV node.

muscle
action
sequence

These depolarizations and repolarizations generate external action potentials which can be recorded at the surface of the body as covered in detail in Chapter 1. These external action potentials generated from within the heart are known as the electrocardiogram or ECG. It is common, also, to refer to this waveform as the EKG, derived from the German spelling of electrocardiogram. The ECG waveform (Fig. 2-5) is shown in relationship to the mechanical action of the heart and the resultant arterial pressure.

heart
electrical
activity
ECG

Electrical activity of the heart is, as stated earlier, initiated by depolarization of the SA node and a resulting contraction of the muscles surrounding the atria. The resulting external action potential is known as the P wave. Immediately following this depolarization, repolarization of the atria occurs; however, for some reason, this does not generate a pronounced action potential. This potential is known as the TA wave and is rarely observed in practice. Electrical activity produced by depolarization of the SA node travels through fibers within the atrium to the AV node. The time taken for this electrical stimulation to travel from the SA node to the AV node is known as the atrioventricular conduction time and is typically between 120 ms and 220 ms. When this stimulation reaches the AV node, this node depolarizes and the depolarization is conducted down through the bundle of His to the myocardium muscle causing ventricular depolarization. The external action potential is referred to as the QRS complex. Immediately following this depolarization the cells concerned repolarize which results in ventricular repolarization or the T wave. Many ECG waveforms also show an additional wave occurring after the T wave. This is designated the U wave and its origin is unknown.

optimum
electrode
positions

The ECG waveform is recorded at maximum potential when one electrode is placed slightly above the heart and to the right and the other electrode is placed slightly below the heart and to the left; thus, the potential output of the heart can be said to be generated along this axis. Chapter 5 discusses variations in the ECG waveform with different electrode positions.

3

MUSCLE ACTION — AND THE SENSORY SYSTEM

Fundamental relationships exist between muscle function, the sense receptors, the brain, the nervous system and the peripheral nerves. The discussion of muscle function that follows includes a discussion of the physiology of the nervous system and the peripheral nerves. The brain is, at this stage, regarded as a processor and is considered in more detail in Chapter 4.

3.1 THE MOTOR UNIT

The motor unit, as the name implies, is the biological unit of muscle function. A motor unit consists of a motor nerve arising from motoneurons in the brain-stem or spinal cord and branching into various motor end-plates. These motor end-plates are each connected to an individual muscle fiber; stimulation of these motor end-plates causes contraction of the single muscle fiber attached to it, as shown in Fig. 3-1. The number of motor units varies between the different muscles of the body;

motor
end-plates

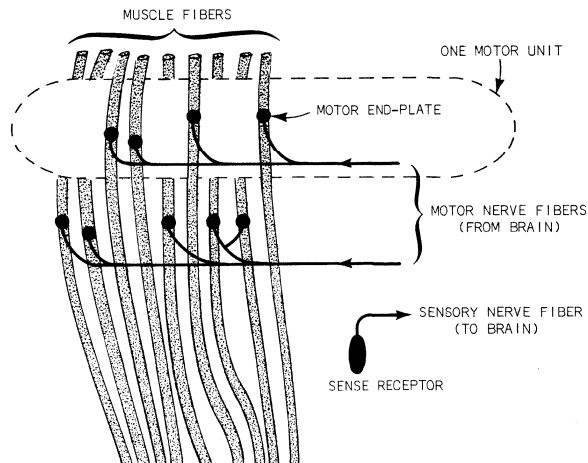


Fig. 3-1. Relationship between nerve and muscle.

generally speaking, the larger the muscle, the more motor units will be found in that muscle. The size of the motor unit, that is, the number of muscle fibers activated by the same nerve fiber, can be quite different for different muscles; in man, one motor unit may contain from 25 to 2000 muscle fibers. The force developed by a motor unit may range from 0.1 to 250 grams weight. The muscle fibers of a motor unit are not clumped together in one part of the muscle, but rather the muscle fibers of different units are interlaced as shown in Fig. 3-1.

3.2 MUSCLE ACTION

movement
smoothness

Chapter 1 stated that a cell can only exist in its polarized or depolarized state; that is, it is a bistable device and intermediate potential levels are not stable. Motor nerves are also cellular in nature, thus, any individual motor nerve can only exist in a polarized or depolarized state and will transmit only two potential levels to the motor end-plates causing a bistable "on-off" action of the muscle fibers. Thus, the individual muscle fibers of one motor unit can exist in only two states, a relaxed state and a tensed state. Normal muscular activity is characterized by smoothness of movement, steadiness and precision. These characteristics are due to the large number of motor units comprising any one muscle. If a small muscular effort is required, only one motor unit will be called into action; as increasing muscular effort is required, many more motor units are called into action until the muscle is providing maximum effort at which time all motor units connecting to this muscle are being used. In this way, some smoothness of movement is obtained.

muscle
modulation

Additional smoothness of movement is obtained by modulating the number of muscle fiber contractions per unit time. Although an individual motor unit can result in only one level of muscular contraction, the number of times that this contraction occurs per unit time (the number of depolarizations and repolarizations executed by the motor end-plate cells) will effectively increase the power of these muscle fibers. Thus, the smoothness of movement of a muscle is controlled both by the number of motor units activated and by the rate at which these motor units are being activated.

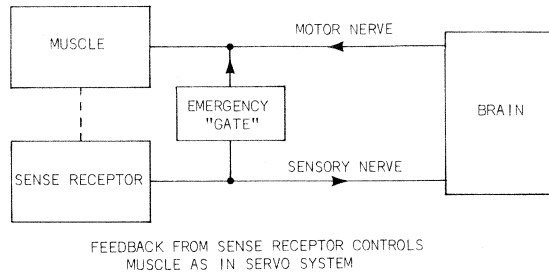


Fig. 3-2. Block diagram of the nervous system.

3.3 THE MUSCULAR SERVO-MECHANISM

A highly simplified block diagram of the nervous system controlling muscle action is shown in Fig. 3-2. The system is similar to a servo-mechanism control system: A sense receptor or transducer produces a position or velocity signal which is sent to the brain via the sensory nerve. The brain in turn initiates an "error" or control signal by comparing the measured position with the desired one stored in the memory. This signal is sent via the motor nerve to the muscle to control its action.

brain
control

This servo-system can be demonstrated by the following experiment. When one's finger is placed onto a cool object, the sense receptor in the finger senses the temperature and relays this information to the brain. The brain interprets this signal as coming from a cool object and, thus, does not necessarily initiate any signal to the motor nerves. If the finger is then placed on a warm object, the brain will interpret the information received from the sensory nerves as relating to a warm object and will activate the motor nerves controlling the muscles in the arms and the hand causing the finger to be lifted from the warm object. There is a time delay of several hundred milliseconds between the time that the sense receptor feels the warm object and the time that the finger is lifted from the warm object. This delay is governed largely by the degree of attention that the subject is paying to the warm object. Now, if the finger is placed on a hot object, a reflex response is obtained and the finger is removed from the hot object in about 150 ms. This reflex is active at all times although it is most marked for the hot object.

150 ms
reflex
response

3.4 REFLEX RESPONSE

emergency gate

An emergency gate has been shown in Fig. 3-2; this emergency gate is not markedly involved in normal receptor/muscle operation. When a reflex response is called for, however, this emergency gate bypasses the signal path to and from the brain and initiates a reflex response. This emergency gate is usually located within the spinal cord. Thus, a reflex response results from a "large" signal, i.e., a high repetition rate signal, being received from a sense receptor. This signal bypasses the brain in the initiation of muscle action. These reflex responses protect the body from serious damage.

3.5 THE POTENTIAL GENERATED DURING MUSCLE ACTION

Depolarization initiated within a sense receptor travels to the brain along the sensory nerve fiber as a series of traveling depolarization waves. The brain then initiates another series of traveling waves of depolarization along the motor nerves to cause a series of depolarizations of the motor end-plates. Depolarization of the motor end-plates depolarizes cells within the muscle fiber causing contraction of these fibers. The actual internal cell potentials involved here are the normal cell polarized potential of -90 mV and the normal cell depolarized potential of $+20$ mV.

recording
action
potential

In dealing with muscles and nerves, it is unusual to use microelectrodes to record the action potential within individual cells. More commonly, needle electrodes are used to record the net result of a number of cells such as one motor unit, or surface electrodes are used to record the results of many motor units. If a microelectrode were inserted into a muscle cell to observe the depolarization and repolarization process, the total process would occur in less than one millisecond. If, however, a needle electrode is placed near these muscle cells, it will detect current flow from many fibers of the corresponding motor unit. These fibers are being fired at their motor end-plates at practically the same instant by the branching nerve. The different fibers of a motor unit do not, however, develop their action currents simultaneously; small time variations between the different fibers occur. These

varying delays are due to the varying lengths of the terminal branches between the motor nerves and the muscle. Thus, the excitation as it travels along is slightly ahead in some fibers compared with others. The result is that current flow in any small area from the cells comprising one motor unit lasts from 2 to 5 milliseconds, which is several times the duration of the current from any single muscle fiber. This asynchronous action helps produce smoothness of muscle action.

With anything other than microelectrodes, the basic electrical activity detected in any muscle is the single motor-unit action potential. These potentials from single motor units are best detected by concentric-needle electrodes or by insulated-needle electrodes as shown in Fig. 3-3. Single motor unit activity may sometimes be detected by placing a small electrode on the skin over the muscle. The volume of muscle influencing such an electrode is, however, considerable and the actual recording is typically the total activity from many motor units having random relationships to one another. The potential produced by muscle action, whether recorded by needle electrodes or by surface electrodes, is known as the electromyogram or the EMG. Measurement techniques for electromyographic recordings are covered in Chapter 12.

EMG

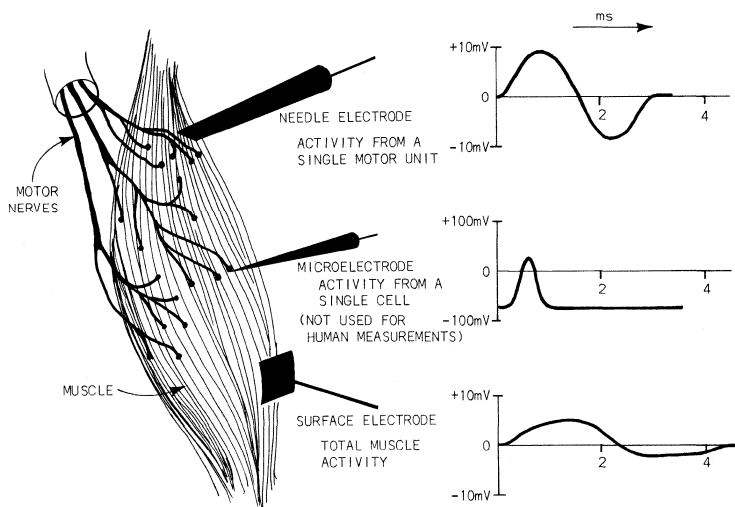


Fig. 3-3. The EMG obtained with various electrode types.

3.6 THE SENSE RECEPTORS

logarithmic
response

If a sense receptor cell or "transducer cell" is activated by a physical distortion, a temperature change, or some other effect, a series of nerve impulses are produced. The time between the individual nerve impulses become greater and greater as various factors come into play. After a while, despite continuation of the physical stimulus, no nerve impulses are generated. It is believed that the sense receptors generate discrete potential levels which are more or less logarithmically related to the stimulus strength. These potentials in turn produce a series of nerve impulses in the nerve fibers. Impulse repetition frequencies from sensory elements vary from a few impulses per second to about 1000 per second; rates above 50 per second are most unusual in human subjects. Since the sense receptors generate an approximate logarithmic response to the stimulus, they can respond over an enormous range ($1:10^7$ or more) of stimulus energy levels.

3.7 THE POTENTIAL GENERATED BY SENSE RECEPTOR STIMULATION

detecting
sense
receptor
activity

It is almost impossible to detect the electrical activity associated with most of the sense receptors on the body; they are small and are not located in clusters of sufficient size to allow detection of the electrical activity associated with a group of receptors. Most sense receptors are not electrical in nature; we measure the electrical activity associated with them only because we do not have the techniques available to us which enable us to observe underlying biochemical mechanisms. There are two groups of sensors that do produce detectable electrical activity: The middle ear and the retina. Potentials associated with the hearing mechanism and the sight mechanism can be detected in the middle ear and on the retina, respectively, because a large number of sensory cells are packed closely together and can be stimulated simultaneously. Only by such an arrangement will enough depolarization current be produced to make detection possible.

ear
response

When a sound reaches the ear, it causes vibration of a membrane within the inner ear known as the basilar membrane. Vibration of this membrane in turn stimulates a large number of sense receptors known as hair cells. If the sound reaching the ear is fairly loud, a large number of these hair cells will develop action potentials at the same time and the current produced will be strong enough to be detected in the middle ear. Electrical potentials precisely following the shape of the stimulating sound waves can be detected by an electrode in the middle ear. This part of the inner ear acts as a transducer and generates electrical activity from sound.

electro-
retinogram

If a bright light is projected onto a substantial area of the retina, many light-sensitive cells within the retina will be stimulated simultaneously and will develop a considerable synchronous response which can be detected from the outside of the retina. The electroretinogram (the external electrical response to light stimulation) may be detected by an electrode consisting of a small, flat silver plate fitted to the inner surface of a small contact lens as discussed in Chapter 11, Section 11.8.

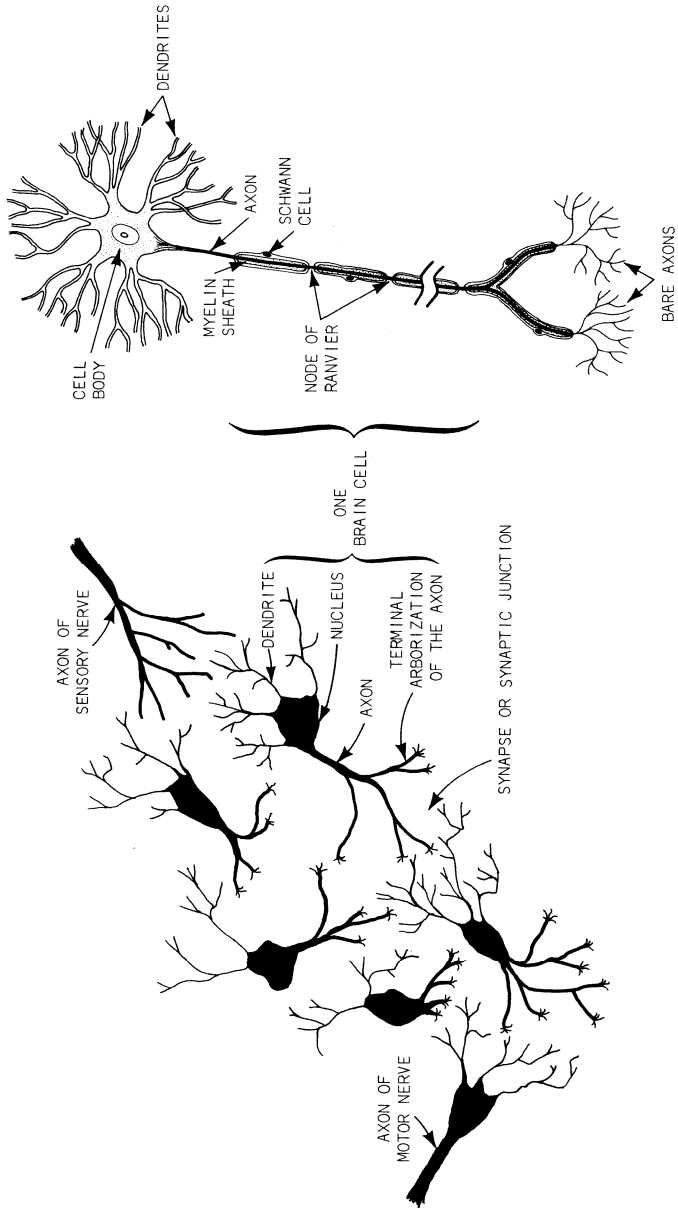


Fig. 4-1. Brain cell interconnections — highly simplified.

4

THE BRAIN AND THE CENTRAL NERVOUS SYSTEM

Chapter 3 discussed sense receptors, muscle action and the potentials associated with the peripheral motor and sensory nerves. To simplify the presentation, Chapter 3 referred to motor and sensory nerves as connections to the brain; strictly speaking, however, most of these nerves connect to spinal nerves in the spinal cord and these *spinal* nerves carry information to and from the brain.

4.1 NERVE CELLS IN THE CENTRAL NERVOUS SYSTEM AND IN THE BRAIN

cell
components

Although nerve cells are located throughout the body and carry information between all body locations, most of the nerve cells are located in the brain and in the spinal cord. The cells comprising the spinal cord and the brain are similar; the cell body is known as the soma and surrounds a region known as the nucleus. The cell body has an extension known as the axon and this may comprise part of the spinal cord and interconnect with sensory and motor nerves or may branch into many small fibers known as the terminal arborizations of the axon. Brain cells consist also of projections of the cell body known as the dendrite which sense information from adjoining cells. The terminal arborizations transmit information to adjoining cells either directly in spinal cells or via the dendrite in brain cells. These cell interconnections (synaptic junctions or the synapse) allow electrical impulses to flow throughout the brain and the central nervous system; one cell acting as a trigger to influence neighboring cells. See Fig. 4-1.

The mechanism by which information is carried is exceedingly complex. As we saw in Chapter 1, the migration of potassium and sodium ions is a primary factor. However, there are many other chemical interchanges involved which cannot be discussed in this text which is specifically concerned with electrical phenomena. It must be emphasized, however, that we measure the electrical activity only because of its *convenience*; it is *not* the fundamental form of nervous activity.

4.2 THE BRAIN

Only a very cursory treatment of the brain is given in this chapter. It is a complex structure, comprised of very large numbers of nerve cells which are interconnected among themselves and which also receive data from the various sensory organs. In a loose sense it can be called a supervisory control unit for it can override most reflexes and direct the body to act in a coordinated manner. Until recently it was thought that most of the functions of the brain resulted from the interaction of nervous impulses at the synapses of the cells (as discussed in Section 4.3) and many analogies with digital electronic computers were formulated. It is now known that this is only part of the story. For example, the substance in which the cells are embodied, the neuroglia (from a Greek word meaning glue), is known to play an important part in brain function and may be vitally concerned with memory. It is thought that oscillatory electrochemical mechanisms are responsible for short term memory but that these impulses gradually modify the chemical structures of parts of the brain so that the permanent storage of information is chemical rather than electrical.

Anatomically, the brain is divided into several parts with the general appearance shown in Fig. 4-2. When sectioned along the midline, the appearance is as in Fig. 4-3. Speaking generally we may say that the deeper structures of the brain, that is those nearest the spinal cord, are those which evolved first and are largely responsible for the more primitive and less easily controlled parts of human behavior (anger, fear, etc.). The cortex (outer structure) is the part of the brain which is highly developed in man and which has enabled him to dominate all other species.

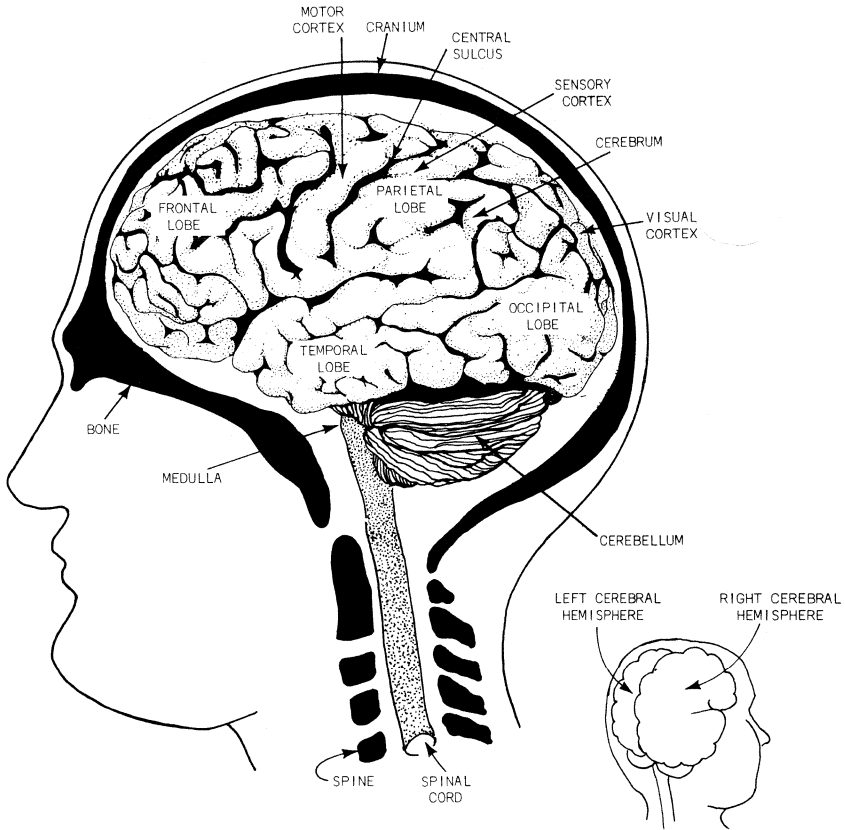


Fig. 4-2. Anatomical sections of the brain.

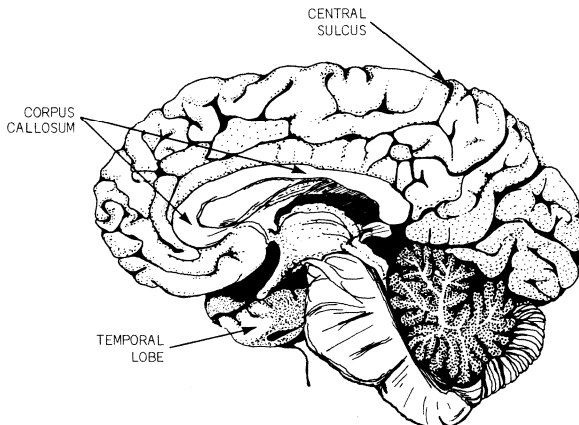


Fig. 4-3. Section of the brain.

Some areas of the cortex serve specific parts of the body. For example, in Fig. 4-2, sensory input is handled in the area marked sensory cortex while motor output proceeds from the motor cortex. Likewise, vision is handled at the rear of the brain in the visual cortex. It is less easy to define areas which serve intellectual functions although the frontal areas are at least in part responsible. There is a large bundle of fibers which interconnect the left and right hemispheres; this is known as the corpus callosum and its location can be seen in Fig. 4-3.

4.3 EXCITATION AND INHIBITION POTENTIALS

The following discussion is a simplified analysis of excitation and inhibition from one scientific school of thought. The properties of excitation and inhibition are, however, highly conjectual and, as stated previously, tell only part of the story. There is no general scientific agreement on the subject.

cell
potentials

In Chapter 1, cells were referred to as bistable devices existing in only two states; a polarized state of -90 millivolts and a depolarized state of +20 millivolts. Intermediate potentials were regarded as having little significance. Nerve cells in the central nervous system and the brain exist in a polarized state of between -70 millivolts and -110 millivolts and the exact cell potential within this range appears to be significant as it determines the cell's vulnerability to a stimulus that would result in regenerative breakdown.

inhibitory effect

excitatory effect

frequency is a parameter

Consider a group of cells producing zero activity; all cells within this group would rest at the normal polarized level of -90 millivolts. Assuming that one cell is depolarized from an external stimulus, this depolarization will affect adjoining cells via synapses, causing a change in the resting potential of these adjoining cells. The effect on these adjoining cells can be either to raise or to lower their resting potential depending on the type of fiber involved in the synaptic junction. If the resting potential is increased, breakdown of the cell membrane will not occur and the impulse can be thought of as having had an inhibitory effect. If the impulse arriving at this synapse decreases the resting potential, the cell is then more susceptible to depolarization and the impulse can be thought of as having had an excitatory effect. With repeated excitatory stimuli at a sufficiently high rate, the membrane will eventually fully depolarize and a new impulse will be propagated. Any single cell is influenced by synapses from many other cells; some are excitatory and some inhibitory. The play of these two opposite effects on any area may or may not cause depolarization of the cell; the deciding factor is the number of impulses being received per unit time and the balance between excitation and inhibition. Although the effect at a synapse of any subthreshold impulse dies away rapidly, it does not die away immediately. This allows almost coincident impulses from different sources or a rapid rate of impulses from a single source to build up their effect; thus frequency becomes an important parameter. Fig. 4-4 shows the effect of various excitation and inhibition signals on any one individual cell, finally resulting in depolarization of the cell once the -70 millivolt threshold is obtained.

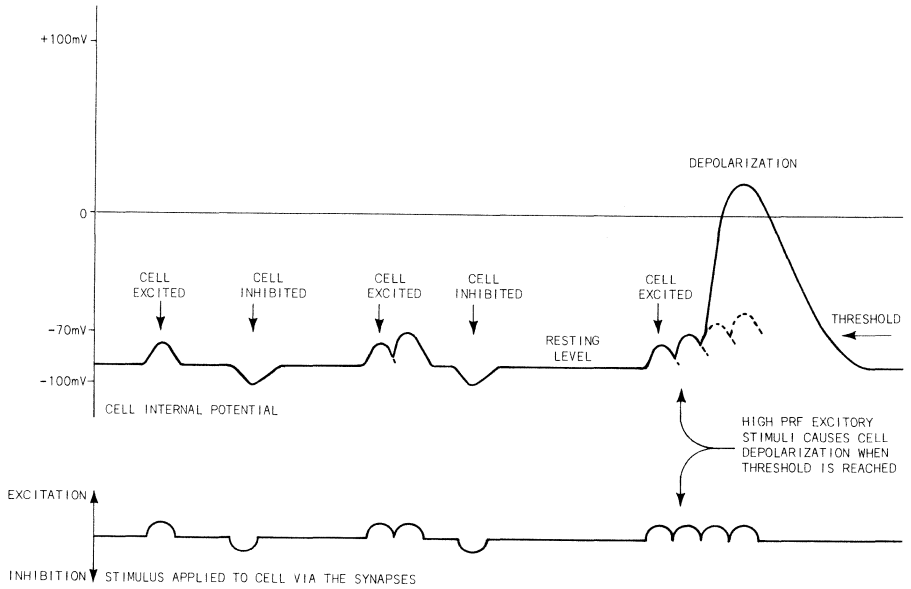


Fig. 4-4. Stimulus changing cell resting potential may result in depolarization of the cell.

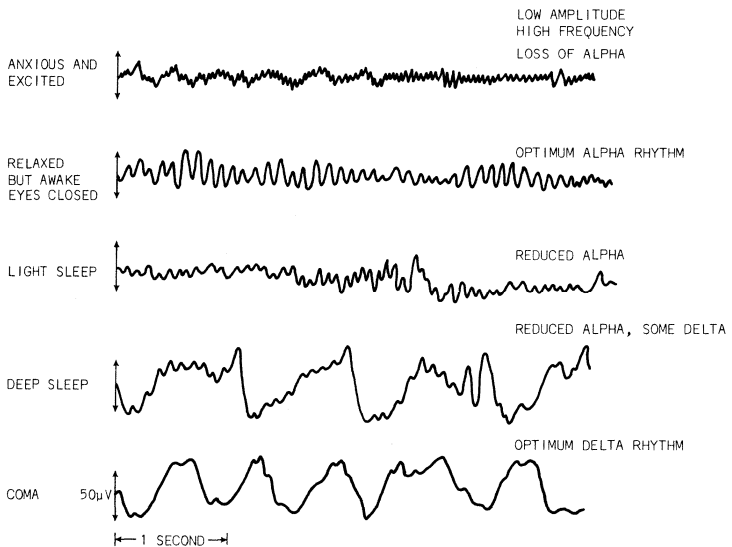


Fig. 4-5. Physiological states and the resultant EEG.

4.4 EVOKED POTENTIALS

producing
evoked
potentials

As shown in Fig. 4-2, parts of the surface of the brain have been found to be associated with various sensory systems. If an appropriate stimulus is applied to a sense receptor, the corresponding sensory area of the brain responds by producing an electric potential known as the evoked potential. The evoked potential, as it appears at the surface of the brain, is the integrative result from the action of many cells. When recorded externally on the scalp, the evoked potential is of the order of 10 μ V or so. It is not yet known completely which cortical elements are responsible for these evoked potentials. Often they are camouflaged by the electroencephalogram, thus it may be necessary to remove the electroencephalogram by an averaging technique when attempting to record the evoked potential. Averaging is discussed in more detail in Chapter 11.

4.5 THE ELECTROENCEPHALOGRAM (EEG)

alpha and
delta
rhythms

As briefly discussed in Section 4.2, the brain can be regarded as a highly developed biochemical factory. The electrical activity which results from so much chemical change is known as the Electroencephalogram (EEG, see Chapter 10) and is in a sense a useful byproduct of nervous action since it allows us to make nonmutilating measurements on an organ which is singularly resentful of external interference. The electroencephalogram, as recorded from the surface of the head, consists of rhythmical, slow sinusoidal waveforms between 10 and 100 microvolts in amplitude. The electroencephalogram varies in both form, amplitude and frequency; the basic frequency of around 10 hertz is known as the alpha rhythm. Should a subject lapse into deep sleep, the alpha rhythm will also disappear and will be replaced by a slower high-amplitude signal known as the delta rhythm. The electroencephalogram produced by a subject in various conditions is shown in Fig. 4-5.

SECTION II

MEASUREMENT TECHNIQUES

The following chapters (5-14) cover some of the more common measurement techniques used in the biophysical sciences. It is intended that this material be used by engineering-oriented personnel as a basis for developing measuring techniques that are uniquely suited to particular requirements. As these requirements are defined by biomedically oriented personnel who will operate the measurement systems, cooperation between the engineer and these personnel is necessary for refinement of the techniques. A working knowledge of electronic measurement techniques is assumed; this material simply applies standard electronic measurement techniques to physiological situations. Typical results are shown for normal subjects and little attempt has been made to define the limits of this normality or to present results that may represent abnormality.

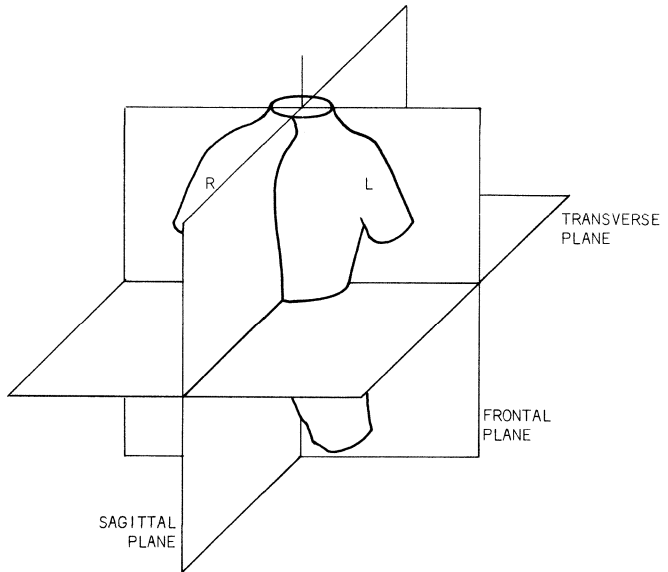


Fig. 5-1. Electrocardiographic planes.

5

ELECTROCARDIOGRAPHY (ECG)

Generation of electrical activity within the heart is discussed in Chapter 2. Electrocardiography is the art of analyzing this electrical activity by measuring potentials at the surface of the body resulting from this electrical activity within the heart. This is achieved by applying electrodes to certain positions on the body and recording the potentials generated between various combinations of these electrodes with an amplifier and CRT display or strip chart recorder.

5.1 ELECTROCARDIOGRAPHIC PLANES

The heart can be regarded as an electrical generator enclosed in a volume conductor, the torso. As this electrical generator is completely enclosed by the torso, a direct measurement of the generator output voltage is impossible without resorting to surgery. The electrocardiographer measures the potential existing between various points on the surface of the volume conductor and uses the information obtained to determine the clinical condition of the heart. The art of electrocardiography is simplified somewhat by considering that at any one time the cardiac potential is projected along axes existing on each of three reference planes: The Frontal plane, the Transverse plane and the Sagittal plane, as shown in Fig. 5-1. This projection of the cardiac potential is referred to as the electrocardiogram (abbreviated ECG or EKG). Various techniques are used to measure the projection of the cardiac potential along axes existing on each of the three planes; these techniques are considered

measuring
cardiac
potential

5.2 FRONTAL PLANE ECG MEASUREMENTS

cardiac
vectors

The electrical potential generated within the heart as projected along axes existing on the frontal plane of the body is known as the frontal-plane cardiac vector, as shown in Fig. 5-2. As with vector measurements in the physical sciences, the relative amplitude and angular position of this vector at any instant cannot be measured with a single measurement. Two separate measurements are required and the results of these two measurements are plotted on a vector diagram to determine the relative amplitude and angular position of the vector. In the physical sciences the two measurements usually determine the projection of the vector along two axes at 90° to one another. In electrocardiography the projection of this vector is recorded along two axes at 60° to one another as it allows the limbs to be used for the attachment of electrodes; the results obtained are relatively independent of where on the limbs the leads are placed.

Einthoven
triangle

Although only two measurements are theoretically necessary to determine the relative amplitude and angular position of the frontal plane vector, it is common electrocardiographic practice to record at least three projections of the frontal plane vector along three axes at 60° to each other. The triangle formed by these three axes is known as the Einthoven triangle as shown in Fig. 5-2. Willem Einthoven, a Dutch physiologist, pioneered many of the electrocardiographic techniques in use today and during the first quarter of this century he developed this vector approach to electrocardiography. Einthoven's law states that the vector sum of the projections of the frontal plane cardiac vector at any instant onto the three axes of the Einthoven triangle will be zero. This is a well-known physical law; however, Einthoven realized that it did in fact also apply to cardiology. Although the above applies to the cardiac vector at *any* instant, cardiologists are usually only interested in the cardiac vector at the peak of the R wave. Thus when the term cardiac vector is used, it implies "Cardiac vector at the peak of the R wave."

cardiac
vector at
R wave peak

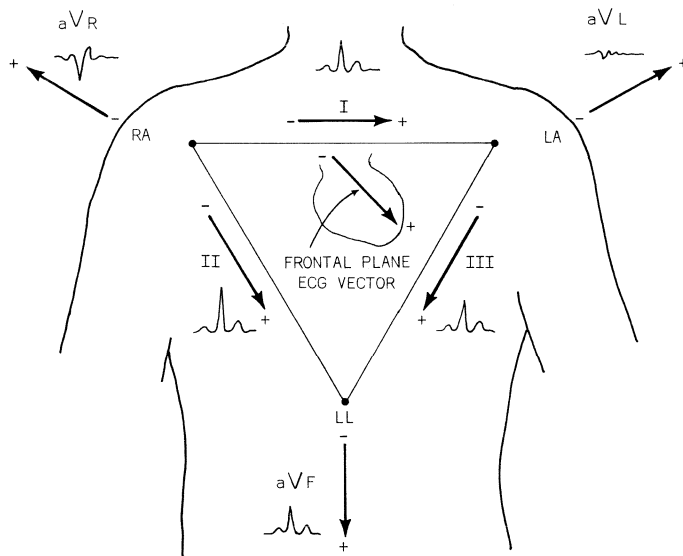


Fig. 5-2. The Einthoven triangle.

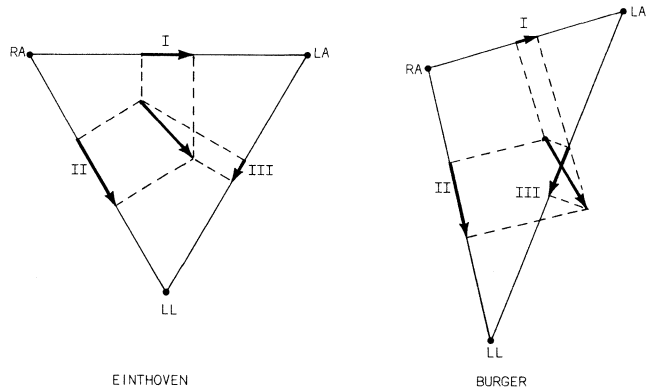


Fig. 5-3. Einthoven's triangle assuming a homogeneous torso, and Burger's triangle with the effects of the lungs and spine.

Burger's
distorted
triangle

More recent advances in electrocardiography have pointed out the well known, but previously ignored, fact that the human torso is neither homogeneous nor triangular and that this leads to a distortion of the electrical field. Thus, the angle of the vector deduced from Einthoven's triangle would be in error. Burger attempted to allow for the inhomogeneities of the torso by introduction of the distorted triangle shown in Fig. 5-3 which compensates for the effect of the lungs and the spine. Although Burger's triangle is perhaps a more accurate representation of the frontal-plane cardiac vector, it is not widely used and cardiologists still prefer Einthoven's approach. They realize that Einthoven's vectors are only an approximation but use them largely for convenience. The clinical interpretation of ECG's is quite empirical in practice, being done by reference to the enormous number of records which have been correlated with known cardiac disorders, usually at autopsy.

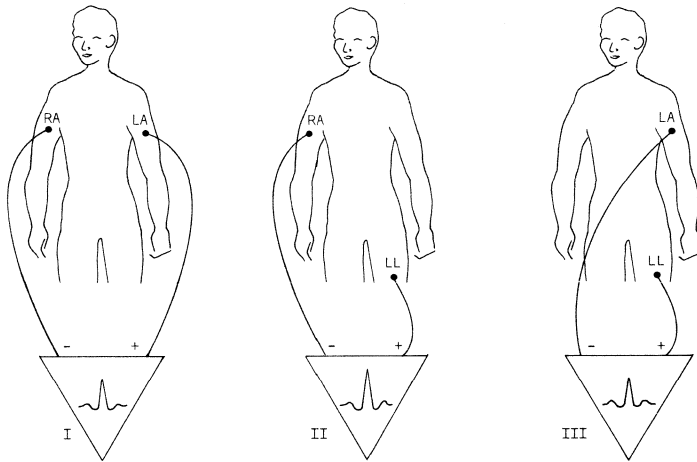
5.3 BIPOLAR LIMB LEAD FRONTAL-PLANE ECG MEASUREMENTS

The three potential measurements commonly used to determine the frontal plane vector in conjunction with the Einthoven triangle are:

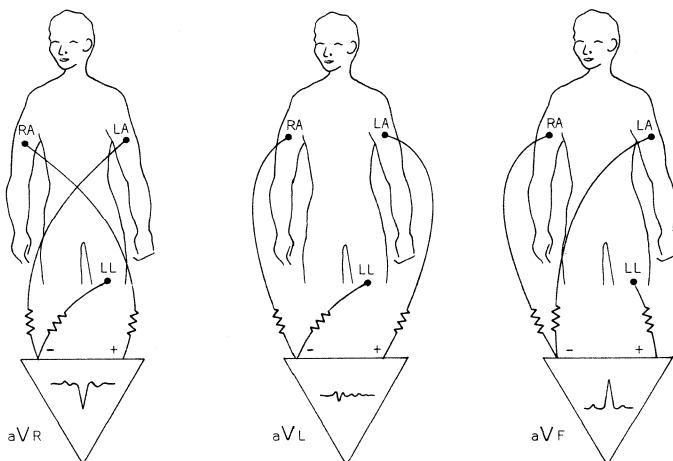
1. Potential between the right arm and left arm.
2. Potential between the right arm and left leg.

3. Potential between the left arm and left leg.

These three ECG measurements are known as the standard frontal-plane or bipolar limb lead measurements and are more commonly referred to as Lead I ECG, Lead II ECG, and Lead III ECG as shown in Fig. 5-4.



STANDARD BIPOLAR LIMB LEADS



UNIPOLAR LIMB LEADS

Fig. 5-4. Electrode positions — frontal plane ECG.

5.4 UNIPOLAR LIMB LEAD FRONTAL-PLANE ECG MEASUREMENTS

augmented
vectors

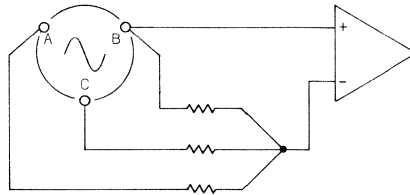
The unipolar limb-lead ECG measurements are a somewhat recent addition to electrocardiography and permit the unipolar registration of the electrical potential in each of three extremities (the left arm, the right arm and the left leg) by creating an "indifferent electrode" or "central terminal" by combining the signal obtained from the other two extremities. The three unipolar limb-lead measurements are referred to as augmented vector right, augmented vector left and augmented vector foot (abbreviated aVR, aVL and aVF, respectively). Augmented measurements provide the same waveshape but 50% more potential output than the now unused, nonaugmented, unipolar limb-lead measurements, VR, VL and VF. The VL, VR and VF measurements were outdated and replaced by aVL, aVR and aVF when recording equipment sensitive to voltage changes rather than current changes became available, as shown in Fig. 5-5. The unipolar limb-lead electrode configurations are simply a projection of the same frontal-plane cardiac vector onto three different 60° axes rotated 30° from the Einthoven configuration, as shown in Fig. 5-6. The three unipolar leads bear a direct vector relationship to the three bipolar standard limb leads.

$$aVR = -\frac{I + II}{2} \quad aVL = \frac{I - III}{2} \quad aVF = \frac{II + III}{2}$$

aVR, aVL,
aVF

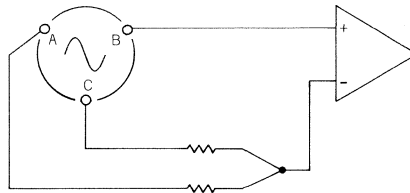
The aVR unipolar measurement refers to the potential at the right arm using the left arm and left leg to form the indifferent electrode. The aVL measurement refers to the potential at the left arm using the right arm and left leg to form the indifferent electrodes and aVF measurement refers to the potential at the left leg using both arms to form an indifferent electrode. It should be noted that in these three measurements the indifferent electrode is formed at the negative input of the amplifier in all cases. The ECG waveform will thus be positive for aVL, positive for aVF, but negative for aVR. Fig. 5-4 shows the electrode positions for recording all frontal plane ECG's.

OLDER, NONAUGMENTED UNIPOLAR CONFIGURATION:



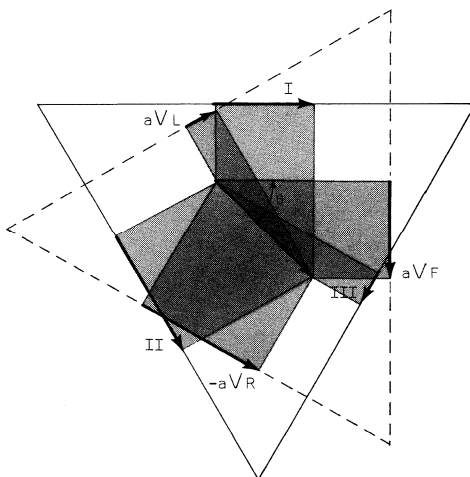
$$\text{POTENTIAL INPUT TO AMPLIFIER} = B - \left(\frac{A + B + C}{3} \right) = \frac{2B - A - C}{3}$$

AUGMENTED UNIPOLAR CONFIGURATION:



$$\text{POTENTIAL INPUT TO AMPLIFIER} = B - \left(\frac{A + C}{2} \right) = \frac{2B - A - C}{2} \left\{ \begin{array}{l} \text{SAME ABC RATIO BUT} \\ \text{50\% MORE POTENTIAL} \\ \text{THAN NONAUGMENTED} \end{array} \right.$$

Fig. 5-5. Equivalent circuit of unipolar measurements.



THE ALGEBRAIC SUM OF THE R AND S WAVE FOR EACH OF THE SIX FRONTAL PLANE ELECTRODE CONFIGURATIONS IS TRANSPOSED TO THIS VECTOR DIAGRAM CONSISTING OF TWO 60° TRIANGLES.



ALGEBRAIC SUM IS A - B

Fig. 5-6. Relationship of frontal plane ECG's to the frontal plane axes of the QRS complex.

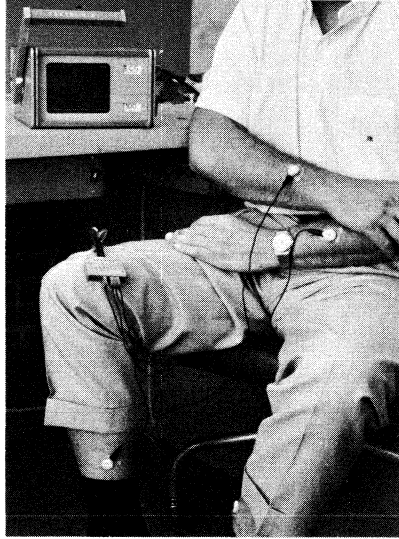


Fig. 5-7. Extremity electrodes: for fully dressed subjects, somewhat “noisy” at times, used in surgery.

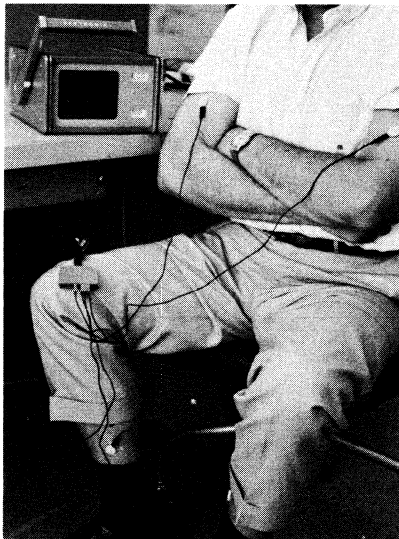


Fig. 5-8. Shoulder electrodes: preferred in most applications, noise free, used in surgery.

5.5 FRONTAL-PLANE AXIS DERIVATION

As stated earlier, only two ECG measurements are necessary to determine the axes of the frontal-plane cardiac vector. It is, however, common to record all six frontal-plane ECG measurement configurations (I, II, III, aVR, aVL, and aVF). A trained electrocardiologist utilizes all six recordings when analyzing the ECG waveform as each ECG recording will differ in shape as well as in amplitude; certain complexes within the ECG waveform are more clearly visible in one configuration than in another.

cardiac
vector
versus
P, QRS and
T waves

To this stage we have simply referred to the electrical output of the heart when recording frontal-plane ECG's as "the frontal plane cardiac vector." This electrical activity is, however, composed of three discrete waveforms as discussed in Chapter 2. These are the P wave, the QRS complex and the T wave. The relative amplitudes of each of these waveforms can be measured separately from the six standard frontal plane ECG measurements and plotted on a vector diagram to determine the electrical axis of a particular segment of the frontal-plane cardiac vector. There is usually a difference of 30° or more between the frontal plane axis of the P wave, the frontal plane axis of the QRS complex and the frontal plane axis of the T wave (see Chapter 6, Vectorcardiography).

5.6 FRONTAL-PLANE ELECTRODE POSITIONING

electrode
placement

Hitherto, the three electrode positions used to record the frontal-plane ECG have been referred to as right arm, left arm and left leg. In actual practice the electrodes can be placed anywhere on the limb or, in fact, anywhere in the general area on the torso near the limb. When recording the ECG of a fully dressed subject, as is usually the case in a doctor's clinic, it is convenient to use the wrists and left ankle as shown in Fig. 5-7. This configuration does restrict movement. Thus, if the subject is to move his body or if the electrodes are to be left in place for several hours, it is more convenient to place the arm electrodes on the upper arms near the shoulders as shown in Fig. 5-8.

If the subject has no upper garment, they are placed on the shoulder blade. Accurate results are also obtained with the electrodes placed in an Einthoven triangle configuration on the chest as shown in Fig. 5-9. This "chest cluster" configuration is used extensively for subject monitoring in intensive care wards and when the subject is exercising during recording. The right leg electrode shown in Fig. 5-7, 5-8 and 5-9 is a ground electrode as discussed in Section 5-9.

"noise" on
waveform

In general, the closer the electrodes are to the heart the better, as any muscle between the electrodes produces "noise" on the ECG waveform. Thus if possible, it is desirable to use the chest configurations to avoid the "noise" generated by the powerful arm muscles. A typical series of frontal-plane ECG measurements is shown in Fig. 5-10.

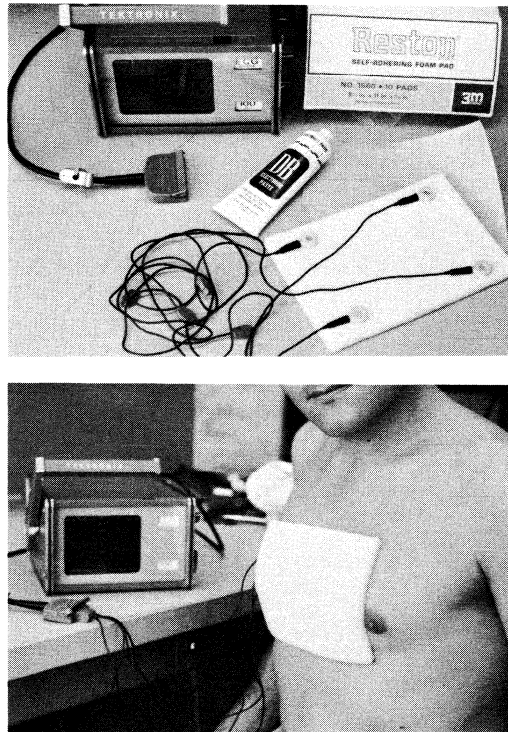


Fig. 5-9. Chest cluster electrodes: for intensive care or exercising subject, difficult to apply in many cases.

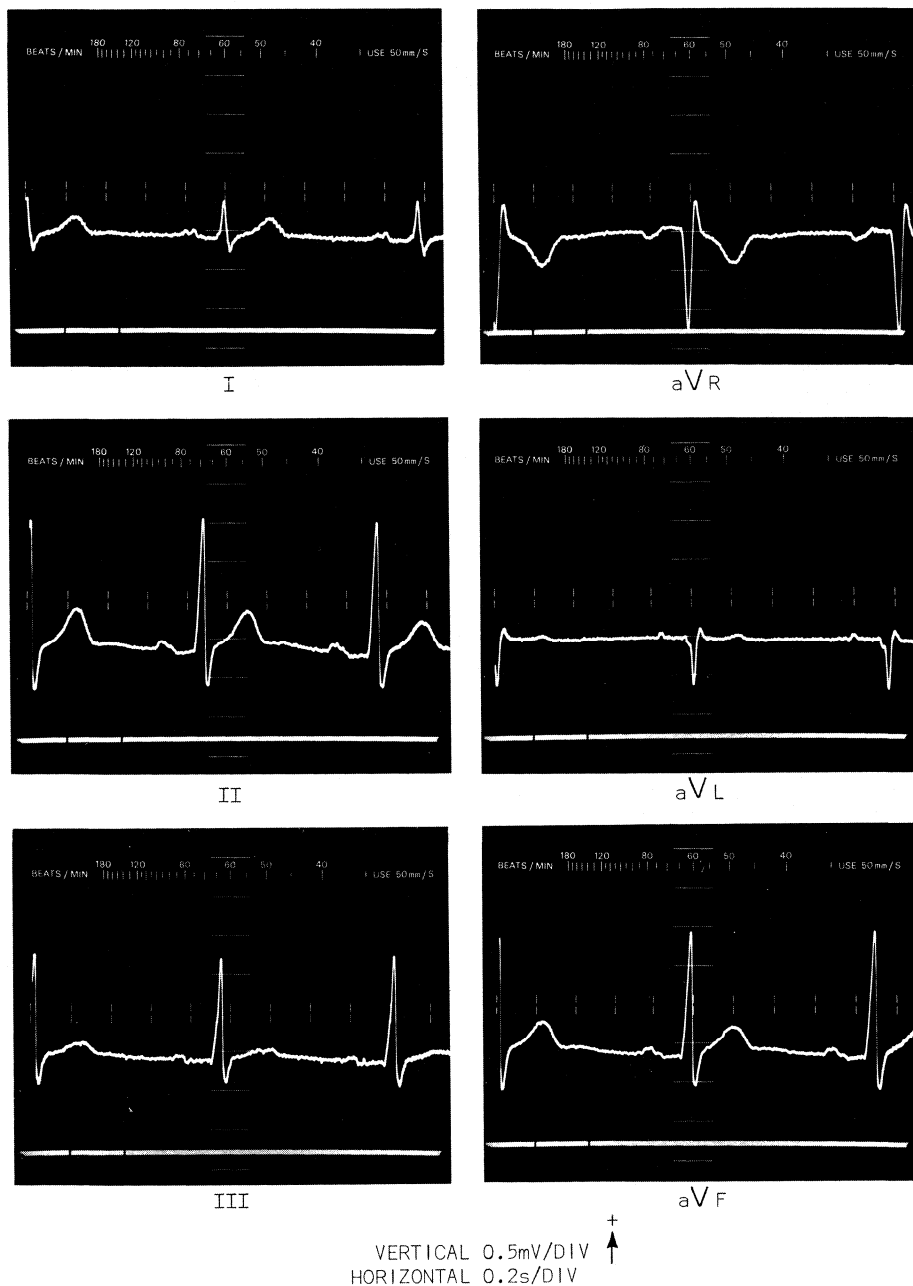


Fig. 5-10. Typical frontal plane ECG's (recorded with Type 410 Physiological Monitor).

5.7 TRANSVERSE-PLANE ECG MEASUREMENTS

electrode
placement

We have previously discussed the measurement of the frontal-plane projection of the cardiac vector; it is common electrocardiographic technique to also record the transverse-plane projection of this cardiac vector. The transverse-plane is shown in Fig. 5-1. Many of the unipolar limb lead measurement techniques discussed earlier apply to transverse-plane electrocardiography. An indifferent electrode is formed by summing the potential at the right arm, the left arm and the left leg and the measurement consists of recording the potential between this indifferent electrode and a chest electrode placed at various positions on the torso. Transverse-plane ECG measurements are known as "the V lead measurements;" V_1 , V_2 , V_3 , V_4 , V_5 and V_6 . The subscripts 1 through 6 refer to the position on the torso of the chest electrode as shown in Fig. 5-11. Although the standard transverse-plane ECG configurations are V_1 to V_6 many other transverse-plane configurations are used. The more common of these other configurations are V_7 , V_8 and V_9 with the chest electrode placed on the left-hand side of the subject towards the back. If the chest electrode is placed in a position somewhat above or below a standard V position, then the configuration is subscripted H or L; thus, if the chest electrode were placed on the midline at the base of the rib cage, it would be referred to as V_{2L} . Occasionally V configurations are recorded with the electrodes on the right-hand side of the subject. Typical transverse-plane ECG's are shown in Fig. 5-12.

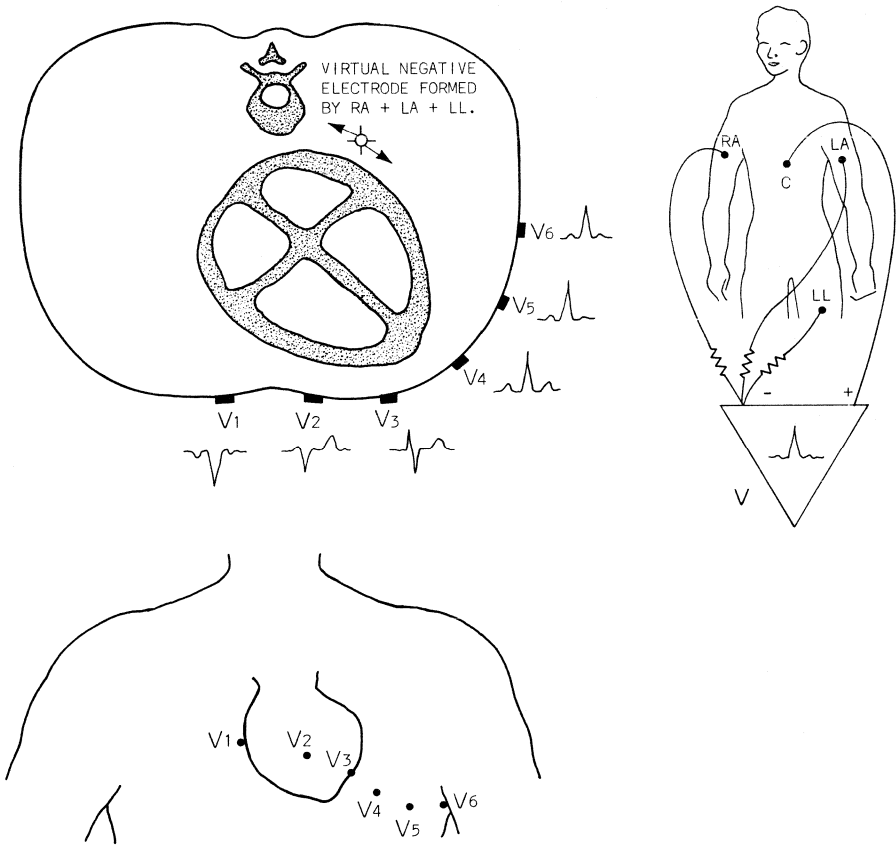


Fig. 5-11. Transverse plane ECG.

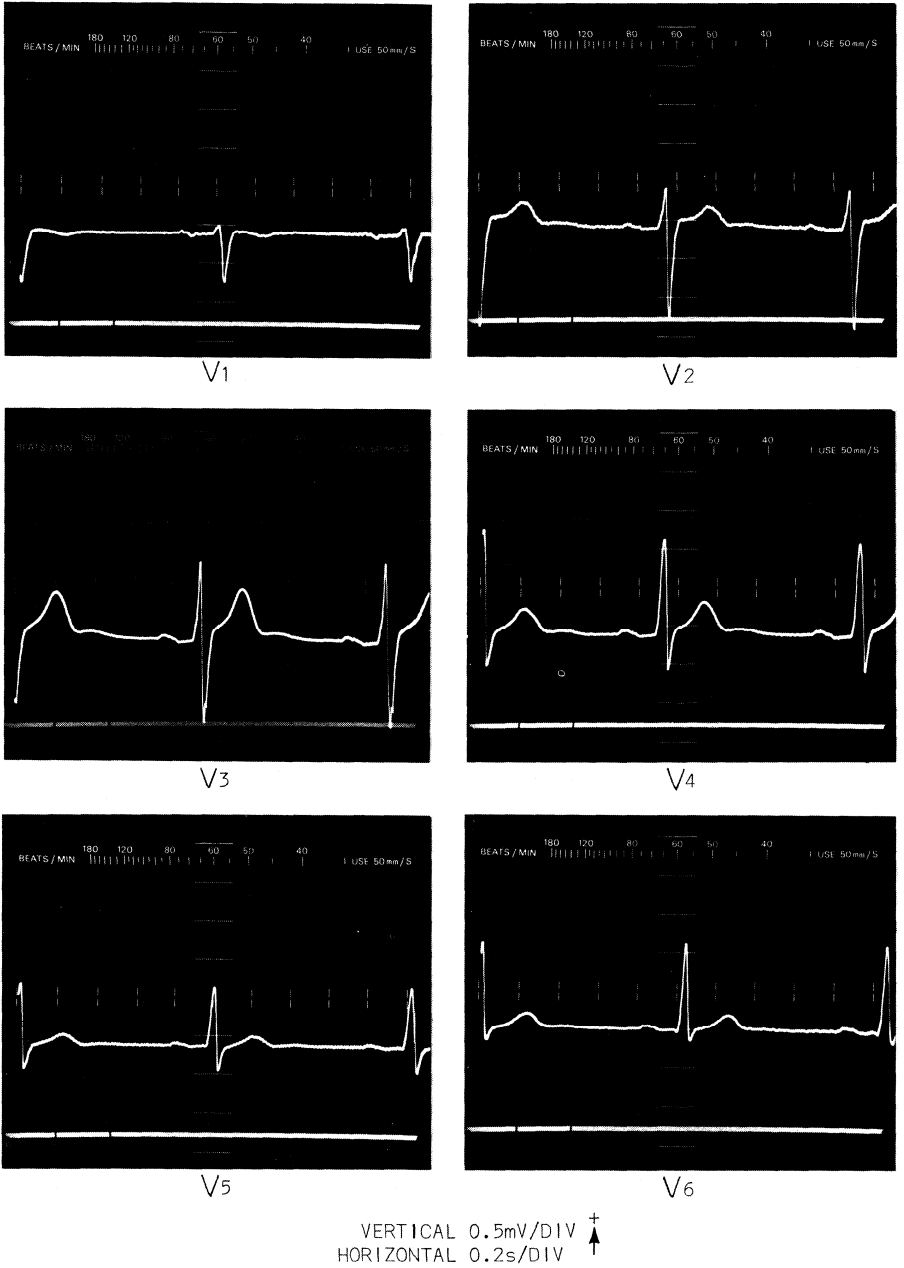


Fig. 5-12. Typical transverse plane ECG's (recorded with Type 410 Physiological Monitor).

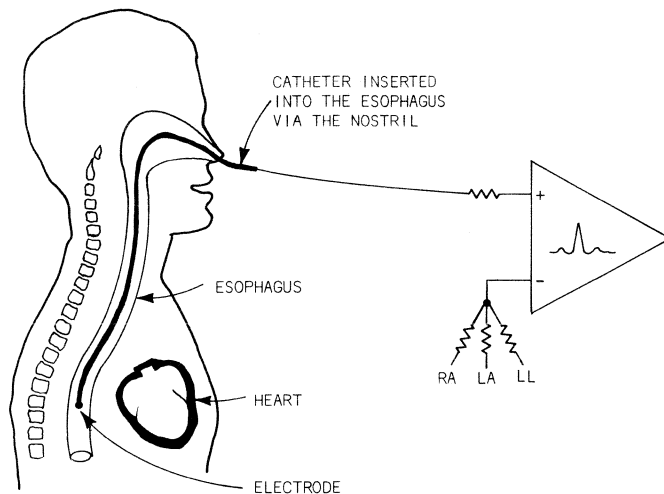


Fig. 5-13. Electrode positions — sagittal plane ECG's.

5.8 SAGITTAL PLANE ECG MEASUREMENTS

The measurements of the cardiac vector projection in the sagittal plane are referred to as "the unipolar esophageal lead ECG measurements" or the "E lead ECG measurements." Although the unipolar esophageal lead ECG is rarely recorded these days, it can be recorded by attaching an esophageal lead to the positive input of the amplifier and forming an indifferent negative electrode using both arms and the left leg as with the transverse-plane V leads. Esophageal lead measurements are recorded from within the esophagus using a catheter, or fine rubber tube, threaded with a wire with an electrode attached to its tip, as shown in Fig. 5-13.

5.9 ECG INSTRUMENTATION REQUIREMENTS

The previous discussion has been concerned with electrode placement and electrode configurations. The potential appearing at these electrodes must be amplified by a differential amplifier and recorded on either a strip chart recorder, a cathode ray tube or, occasionally, magnetic tape.

display
devices

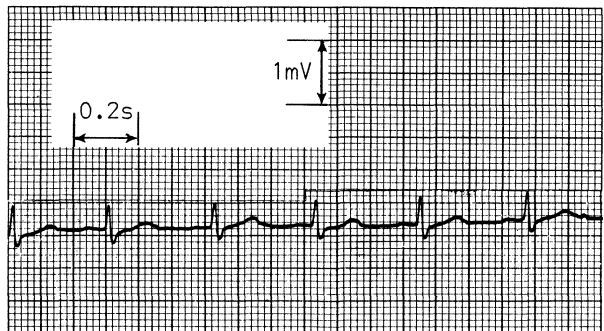
Commercial instruments usually include electrode selection, amplification and either a CRT display or a strip chart recorder. These instruments are commonly referred to as ECG monitors or physiological monitors if they utilize a CRT display and as electrocardiographs if they utilize a strip chart recorder display. The commercial electrocardiographs preceded commercial physiological monitors by 50 years or so; thus, many of the features on physiological monitors are carried over from electrocardiographs even though they may not be optimally suited for use with CRT displays.

electro-
cardiograph

Electrocardiographs almost invariably use graph paper with horizontal and vertical lines at 1 millimeter intervals and a heavier line at 5 millimeter intervals as shown in Fig. 5-14. In routine electrocardiography the recording speed is 25 millimeters per second; thus, the 1 millimeter horizontal intervals each represent 0.04 seconds and the 5 millimeter intervals represent 0.2 seconds. The sensitivity of electrocardiographs is typically 10 millimeters per millivolt.

CRT
displays

When referring to CRT displays, it is normal practice in the electronics industry to refer to millivolts/centimeter vertically and seconds/centimeter horizontally. However, due to the historical influence of electrocardiographs, the reciprocal of these dimensions is used in electrocardiography; i.e., millimeters/millivolt and millimeters/second.



COMPUTER INSTRUMENTS CORP. HEMPSTEAD, NEW YORK RP-120

Fig. 5-14. A typical electrocardiograph tracing.

Physiological monitors commonly use twice the vertical sensitivity and twice the sweep speed of electrocardiographs; that is, 20 millimeters/millivolt and 50 millimeters/second.

right leg
ground
electrode

As high input impedance differential amplifiers are almost invariably used in modern electrocardiographs and physiological monitors, some ground reference must be maintained between the subject and the amplifier. This is usually accomplished by attaching an electrode to the subject's right leg and connecting this electrode to the ground of the amplifier. For reasons covered in Chapter 16 (Electrodes) and Chapter 19 (Amplifiers), amplifiers used to record the ECG should have a common mode input impedance of at least 10 megohm per input and should have a common mode rejection ratio of at least 10,000:1. Frequency components within the ECG extend from .05 Hz to approximately 80 Hz; most commercial amplifiers will exhibit adequate high frequency response, however many amplifiers do not exhibit a low frequency response to .05 Hz or better.

subject
protection

It is highly desirable that any amplifier used to record ECG's, or used to record other potentials from the human body, should incorporate internal circuitry to protect the subject against electrical shock should the amplifier fail. Protection circuits are covered in Chapter 17 (Safety).

5.10 INTERPRETATION OF THE ELECTROCARDIOGRAM

Interpreting the results obtained from the various ECG measurements discussed previously is an art in itself and should only be attempted by suitably trained medical personnel. Although it is not within the scope of this publication to cover ECG interpretation, it is desirable that the reader have some concept of what is considered normal and what is considered abnormal and appreciate the meanings of the more important terms used by cardiologists when discussing the shape of the ECG waveform.

The "normal" ECG has come about by observing the distribution in many thousands of healthy subjects. The ECG's presented in Fig. 5-11 and Fig. 5-14 are "normal" ECG's from a healthy adult male. Any

abnormalities variation from the normal rhythm of an ECG is known as an arrhythmia. As arrhythmias may be more evident in certain ECG measurements, a complete set of frontal-plane and transverse-plane ECG's is usually required for a complete analysis of the electrocardiogram.

A fast heart rate in excess of 100 per minute is known as tachycardia; a slow heart rate of less than 60 per minute is known as bradycardia. An excessively large and continuous ECG with no recognizable P, QRS or T waves is a sign of ventricular fibrillation, a condition where the ventricular muscle goes into local oscillation in a "circus movement," waves of depolarization going round and round the ventricular tissue. A random and changing phase relationship between P and QRS and T indicates adioventricular rhythm, in which a complete bundle block occurs in the bundle of His, and the auricles and ventricles are beating independently, with the ventricles slower. Arrhythmias also appear as extra beats, known as ectopic beats, on an otherwise normal electrocardiogram. A good reference for the study of cardiac arrhythmias is *Principals of Electrocardiography* by J. Goldman or *Coronary Care Unit Nursing, Part I* by H. A. Braun and G. A. Diettert.

6

VECTORCARDIOGRAPHY (VCG)

The electrical activity of the heart as projected along various axes on three planes of reference is referred to as the ECG and is discussed in Chapter 5. Vectorcardiography is the art of analyzing the electrical activity within the heart by obtaining ECG's along three axes at right angles to one another and displaying any two of these ECG's as a vector display on an X-Y oscilloscope. This display is referred to as a vectorcardiogram or VCG. The three axes chosen are designated X, Y, and Z as shown in Fig. 6-1. The planes bounded by any two of these axes are also shown in Fig. 6-1.

derived
from ECG
measurements

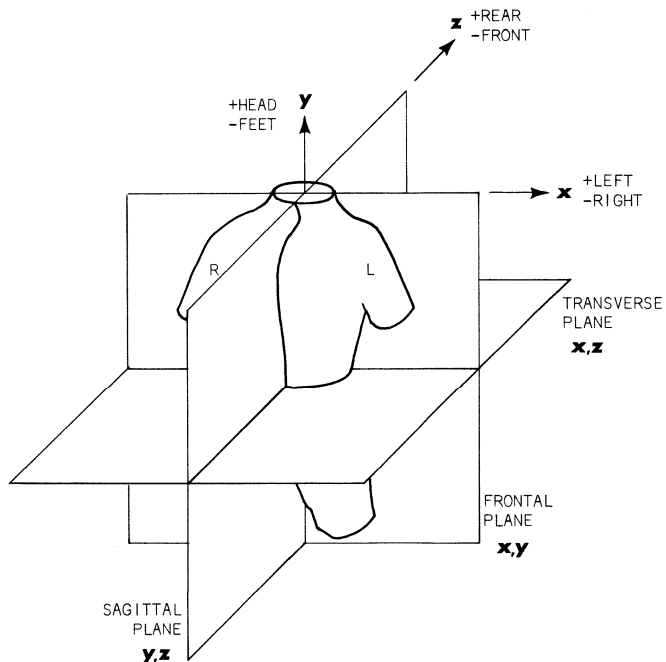


Fig. 6-1. VCG planes.

The electrocardiogram displays the electrical potential in any one single axes, however the vectorcardiogram displays the same electrical events simultaneously in two perpendicular axes. At present, vectorcardiography is used only in medical centers and by specialty medical groups. Its value is limited for routine clinical practice due to the expense of the necessary equipment, the time involved, and because no definite standard as to the best lead system to be used or as to what is "normal" has been established.

6.1 THE SPATIAL VECTORCARDIOGRAM

The spatial vectorcardiogram may be defined as the record of the time variations of the instantaneous vectors which represent the electrical activity of the heart. Fig. 6-2 shows the spatial vectorcardiogram and the resulting projections of this spatial vectorcardiogram onto the frontal plane, the transverse-plane and the sagittal plane. The projections are referred to as the frontal VCG, the transverse VCG and the sagittal VCG.

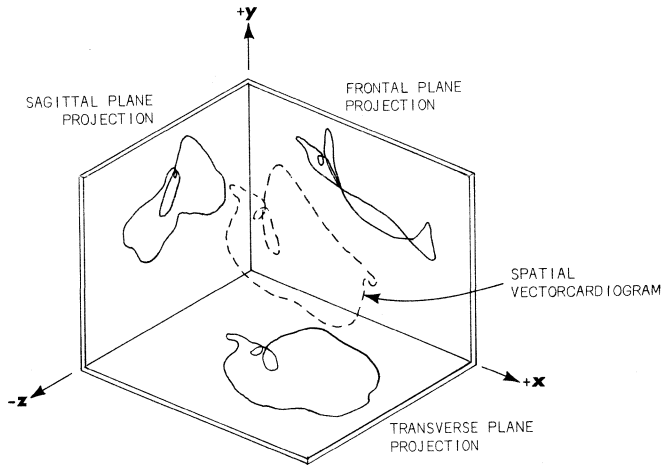


Fig. 6-2. VCG projections.

6.2 ELECTRODE PLACEMENT

The location of electrodes on the body to allow accurate recording of the ECG's along the X, Y and Z axes is extremely important. To obtain an accurate vectorcardiogram, the electrodes must be placed in positions that permit accurate recording of the three vectors. The body and the heart, however, exhibit characteristics which cannot be circumvented for practical purposes. This makes absolutely accurate recording of the vectorcardiogram almost impossible as any system of electrode placement is somewhat imperfect.

absolute
accuracy
difficult

The vectorcardiogram can be recorded by obtaining the potential between two points located under the arms for the X or left to right component, between the chest and the back for the Z component and between the head and left leg for the Y component. Simply displaying any two of these three signals on an X-Y oscilloscope with the same sensitivity for both the X and the Y amplifiers would result in considerable errors as the potential generated within the heart is attenuated by different amounts in the X, Y and Z directions and, as the torso is not homogeneous, these signals are generated along axes which are not exactly at 90° to one another.

6.3 FRANK ELECTRODE SYSTEM

seven
electrodes
compensating
network

Various attempts have been made to compensate for the above attenuation and torso inhomogeneities. The Frank system of vectorcardiography attempts to achieve isotropy as well as orthogonality by using seven critically placed electrodes and a resistive network. The Frank electrode positions and the Frank attenuation and compensation network is shown in Fig. 6-3.

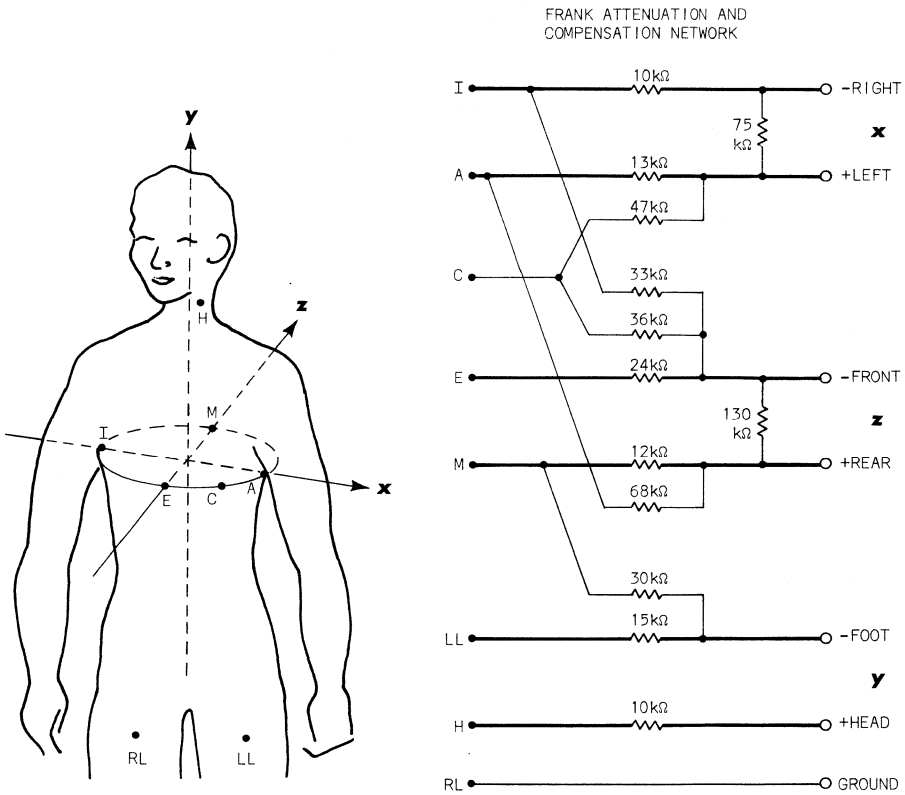
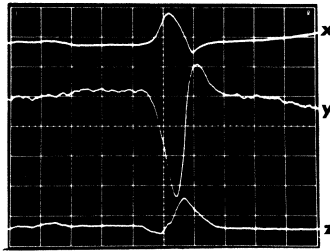


Fig. 6-3. Frank electrode positions.

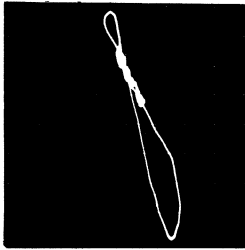


THE **x**, **y** AND **z** OUTPUTS OF THE FRANK ELECTRODE NETWORKS

.04s/DIV
0.5mV/DIV

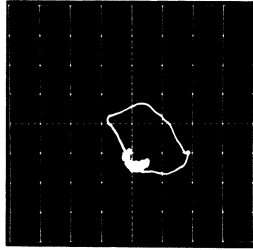
THREE TRACE DISPLAYS SHOWING THE THREE QRS COMPONENTS OF THE VECTOR DISPLAYS SHOWN BELOW.

y VERTICAL
x HORIZONTAL



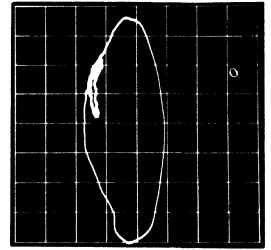
FRONTAL

z VERTICAL
x HORIZONTAL



TRANSVERSE

y VERTICAL
z HORIZONTAL



SAGITTAL

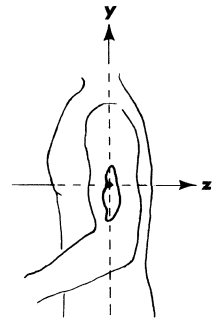
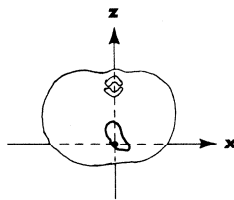
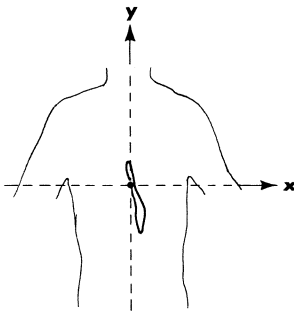


Fig. 6-4. The Frank vectorcardiogram.

6.4 POLARITY CONVENTION

+X to left
side, +Y to
head, +Z to
back

Fig. 6-3 shows a positive direction for X towards the subject's left-hand side, a positive direction for Y towards the head and a positive direction for Z towards the back. Since all modern oscilloscopes adopt the convention of a positive potential change deflecting the beam upwards and towards the right-hand side of the screen, it is necessary to adopt this polarity convention for the three axes to present the three vectorcardiograms superimposed on the human body as viewed from logical viewpoints; i.e., from the front, from the top and from the left-hand side. The polarity of the ECG waveform in the X and Z direction will thus appear "positive," however the polarity of the normal ECG waveform in the Y direction will be inverted, as shown in Fig. 6-4. Frank's original network suggested a positive Y direction towards the feet which resulted in the three separate ECG components being represented as "positive" signals. This resulted in inversion of the frontal and sagittal projections of the vectorcardiogram, making correlation to the actual position of the heart somewhat difficult. The conventions adopted in this chapter are nowadays regarded as standard.

6.5 OTHER ELECTRODE SYSTEMS

Fig. 6-5 shows three other electrode configurations in common use, the axial electrode placement, the tetrahedron electrode placement and the cube electrode placement. Although none of these three configurations are considered as accurate as the Frank system of vectorcardiography, they do not require complicated resistive networks and can utilize two standard ECG monitors and an X-Y oscilloscope to obtain satisfactory results. Indifferent electrodes are formed by summing the effects of various electrodes in the same way as for electrocardiography as covered in Chapter 5.

6.6 THE NORMAL VECTORCARDIOGRAM

QRS loop
predominant

Vectorcardiograms obtained from a healthy, male Caucasian subject using the Frank system are shown in Fig. 6-4. The upper CRT display shown in Fig. 6-4 represents the QRS segment of the three separate ECG's used to form the vectorcardiograms shown in the lower CRT photographs. Careful inspection will reveal that the three QRS segments are not exactly in phase; if they were, the vectorcardiogram loops resulting from the QRS segment would all be straight lines. Although each vectorcardiogram should theoretically exhibit three loops showing the vector orientation of the P wave, the QRS axis and the T wave, the loop obtained from the QRS complex predominates and an increase in horizontal and vertical deflection factors is normally necessary to adequately view the loops resulting from the P wave and T wave. It should be noted that the VCG is very dependent on the phase of the respiratory cycle in which it is recorded. The posture of the subject is also very important.

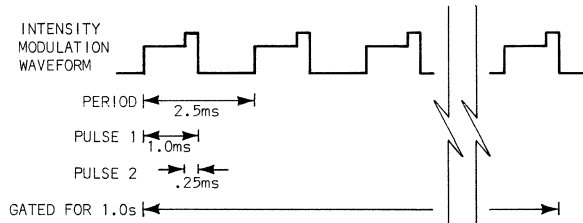
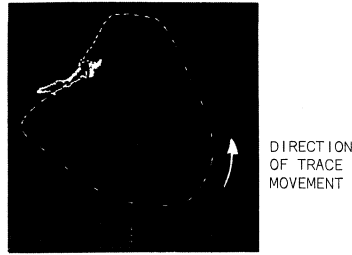


Fig. 6-6. Sagittal VCG with 2.5-ms time-duration markers.

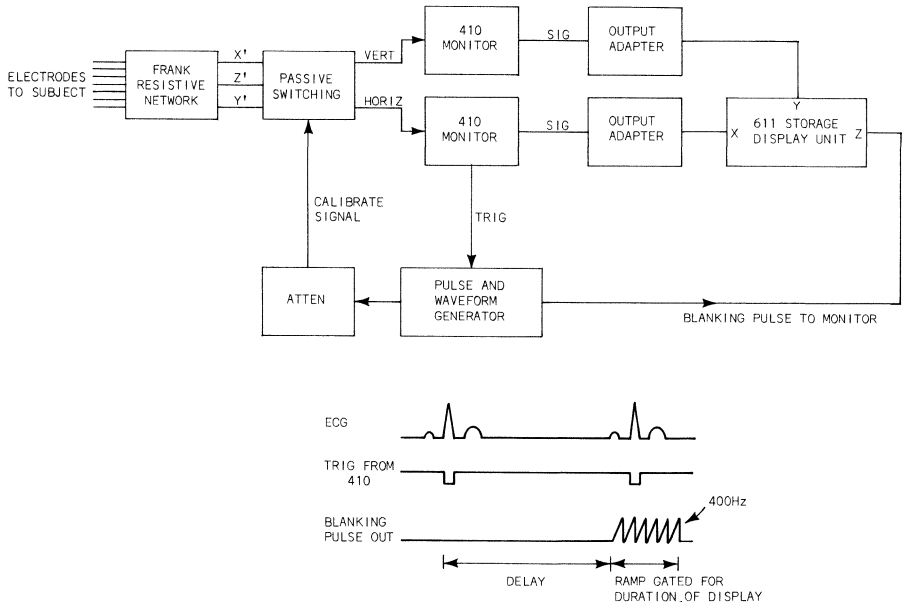


Fig. 6-7. VCG system using two monitors and incorporating intensity modulation.

6.7 TIMING AND DIRECTION REFERENCE

intensity
modulation

The vectorcardiograms shown in Fig. 6-4 give no information as to the direction of movement of the CRT beam when inscribing the vector loops, nor do they provide any indication of the time taken to inscribe the loops. As this information is often desirable, some form of intensity modulation is usually incorporated into VCG instrumentation to provide a time scale and direction reference. The single sagittal vectorcardiogram shown in Fig. 6-6 includes intensity modulation at 400-Hz rate to provide a timing reference of 2.5 milliseconds between each consecutive brightened area on the display. Two discrete intensity modulation levels are used for each intensifying pulse providing an "arrowhead" effect to indicate the direction of movement of the CRT beam when inscribing the vector loops. As with the VCG's shown previously in this chapter, only the QRS loop is clearly displayed. Some VCG systems utilize sawtooth intensity modulation in place of the two-step intensifying pulse.

6.8 INSTRUMENTATION REQUIREMENTS

A complete vectorcardiography system is shown in Fig. 6-7. Many of the ECG instrumentation techniques discussed in Chapter 5 are used when recording the VCG. VCG measurement requires two separate ECG measurement systems as well as a Frank resistive network, an X-Y storage oscilloscope and intensity modulation circuitry.

The circuit of the Frank resistive network is shown in Fig. 6-3. Further details of this network are given in Chapter 29. The passive switching network selects the correct combination of the X, Y and Z signals from this Frank network for displaying either the frontal, transverse or sagittal vectorcardiogram. The two ECG's selected are displayed on individual 410 monitors and the output of each of these monitors is coupled via passive output adapters to the 611 storage display unit.

410 with
611

The output adapters are necessary to decrease the signal output of the 410 to the level required by the 611 and to allow calibration of the system from an external calibration signal. For details of these adapters refer to Chapter 29. A combination of Tektronix pulse and waveform generation modules (Refer to Chapter 22) is triggered by the trigger signal available from either of the 410 monitors to provide blanking pulses to the CRT. This blanking information adds time and direction information to the vector display. These modules also provide a calibration signal which is fed to the passive switching network via an attenuator. A C-10 camera may be used to record the VCG from the 611 display unit.

5031

The Tektronix 5031 storage oscilloscope operated in the X-Y mode is ideally suited to vectorcardiology. The 5031 may be used as a complete vectorcardiograph system by using the cube electrode placement given in Fig. 6-5. Normally the internal time base of the 5031 is disabled when operating in the X-Y mode. However, if this disabling circuit is removed by removing pin Z from the sweep circuit board, the internal sweep circuitry can be triggered from the calibrator and used to provide a multilevel intensity modulation as shown in Fig. 6-6. If desired, a Frank resistive network as detailed in Chapter 29 could be used ahead of the 5031. Any oscilloscope camera may be used to record the results.

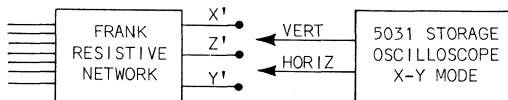


Fig. 6-8. Simplified VCG system.

561B or
564B with
3A74

Many other vectorcardiograph systems are in use. One of the more common alternate techniques is to use a 561B or 564B oscilloscope with 3A74 four-trace units in both the vertical and horizontal channels. When used in conjunction with three preamplifiers, all three projections of the spatial vectorcardiogram can be displayed simultaneously. The display can then be photographed and the photograph enlarged for convenient viewing. Additional information on the components required for these VCG systems are given in various chapters in Section III. The custom instrumentation required for the above systems are discussed in Chapter 29.

410 subject
protection

It should be appreciated that the Tektronix 410 monitor is the only Tektronix product incorporating subject protection features. While other Tektronix products include some protection and are inherently highly reliable, current limiting adapters as described in Chapter 29, Section 29.12, should be used when using these instruments in conjunction with human subjects. Safety is covered further in Chapter 17.

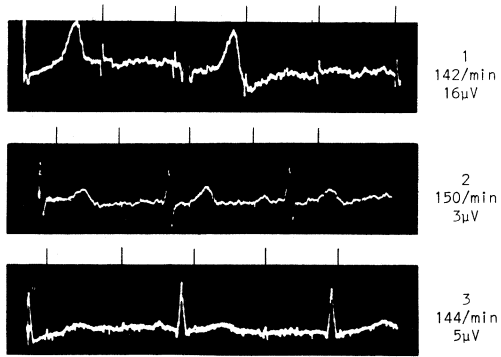


Fig. 7-1. Fetal ECG's at 16 weeks on several subjects.

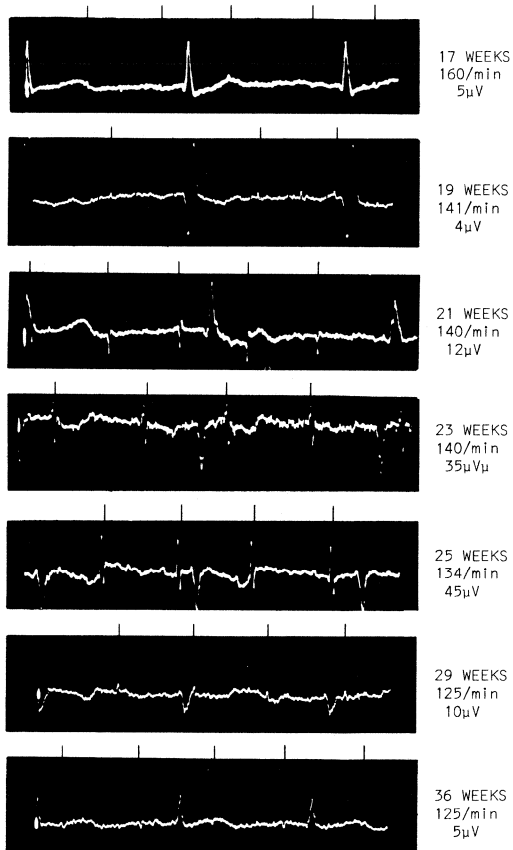


Fig. 7-2. Fetal ECG variations with gestation time.

7

FETAL ELECTROCARDIOGRAPHY

Fetal electrocardiography is the art of recording the ECG of a fetus in utero by placing electrodes on the mother's abdomen and using recording techniques similar to those used for normal electrocardiographic recording. The procedure is complicated by the existence of the mother's ECG and muscle action potentials and by the severe attenuation suffered by the fetal ECG when recorded at the mother's abdomen. The fetal ECG is normally abbreviated F-ECG.

7.1 THE EXISTENCE OF THE FETAL ECG

best from
16 to 24
weeks

The fetal ECG may be recorded as early as the 11th week of gestation in some subjects and can be recorded in almost all cases after 16 weeks gestation. Fig. 7-1 shows the variation in fetal ECG amplitude that may be expected at 16 weeks. By the 18th week the fetal ECG is invariably pronounced and continues to become more evident to about the 24th week. At this stage, amniotic fluids begin to build up in the mother which reduces the amplitude of the fetal ECG. This reduction slowly continues for most of the remaining weeks of pregnancy. Just prior to delivery the amniotic fluids are released and the fetal ECG can again be prominently recorded. The variation in the fetal ECG obtained from a subject at various gestation times is shown in Fig. 7-2.

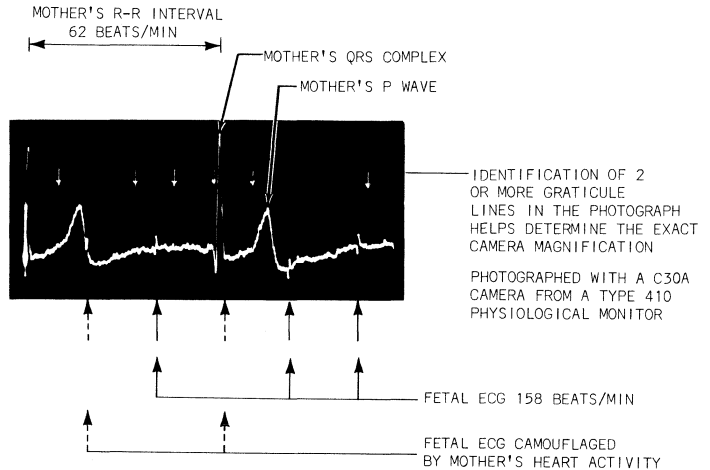


Fig. 7-3. A typical ECG display at 18 weeks.

7.2 THE NORMAL FETAL ECG

As previously stated, the fetal ECG amplitude varies greatly with gestation time and also varies from one subject to another. A typical fetal ECG obtained at 18 weeks is shown in Fig. 7-3. The mother's ECG is clearly evident and, being many times greater in amplitude than the fetal ECG, may camouflage the fetal ECG. The fetal ECG can clearly be observed during the isoelectric period between the mother's ECG. The techniques used to record this fetal ECG are covered late in this chapter. From this recording, it is evident that the mother's heart rate is 62 beats per minute and the fetal heart rate is 158 beats per minute. In the early stages of pregnancy the fetal heart rate is normally between 140 to 160 beats per minute. If the mother is relaxed during the recording of the fetal ECG, it is expected that her heart rate would be under 80 beats per minute. Due to the low amplitude of the fetal ECG, the fetal R wave is normally the only part of the fetal ECG complex that is evident.

7.3 SUBJECT PREPARATION

As the electrical potential generated by muscular activity within the mother's abdomen will clearly be present using abdominal electrodes when attempting

relaxed
state
imperative

to record the fetal ECG, it is imperative that the mother be situated comfortably and in a state of complete rest. A bed or couch is normally preferred to a clinical examination table. To reduce the mother's muscular activity in the abdominal area as much as possible, it is desirable that the subject does not eat for several hours prior to recording and that the subject empty her bladder shortly before recording.

7.4 ELECTRODE PLACEMENT

Many electrode positions have historically been used to record the fetal ECG. It is now generally accepted that abdominal electrodes are preferable to most other configurations. Due to the variation expected in the fetal position, many electrodes should be applied to the abdomen and the fetal ECG recorded between various combinations of these electrodes until a positive recording is obtained. The most common fetal ECG lead configuration is the Blondheim configuration shown in Fig. 7-4. This configuration consists of three electrodes (A, D and F) placed at the vertices of a 60° triangle and an additional three electrodes (B, C and E) placed at the vertices of an inverted 60° triangle. Fig. 7-4 also shows two electrodes on the back of the subject (G and H). It has been found that, during the early stages of pregnancy, a recording from one of these two back electrodes to any of the abdominal electrodes produces more reliable results. It is important that care be exercised when applying the electrodes.

Blondheim
configuration

USE ONE OF THESE 8 ELECTRODES AS "RL" OR GROUND OR AFFIX AN ADDITIONAL RL ELECTRODE ANYWHERE ON THE ABDOMEN.

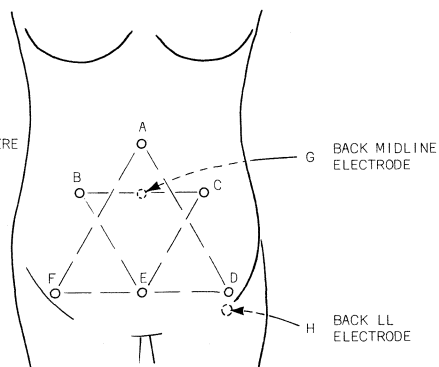
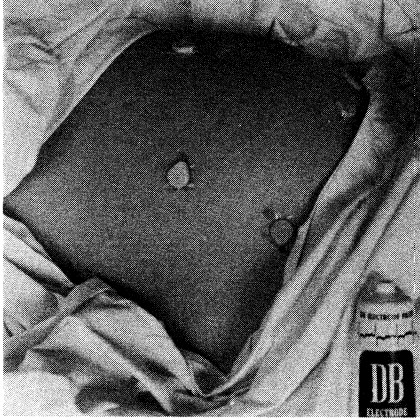
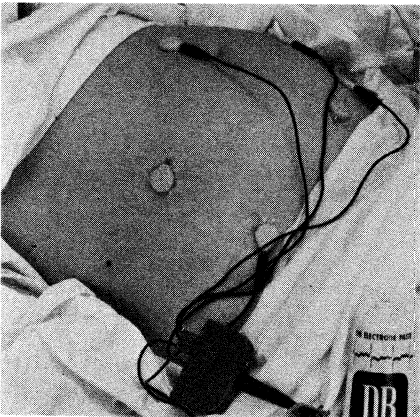
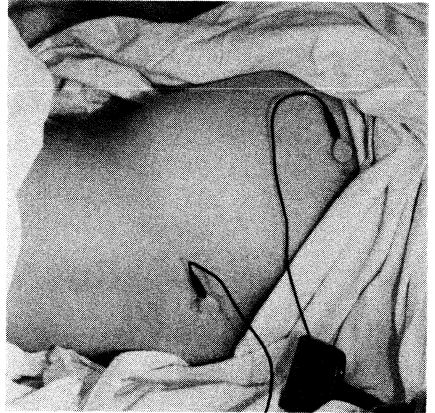


Fig. 7-4. Fetal ECG electrode positions.



ELECTRODES AFFIXED TO ABDOMEN

TWO BACK ELECTRODES AFFIXED AND ADAPTER CABLES FITTED TO THEM



ADAPTER CABLES FITTED TO ABDOMINAL ELECTRODES

Fig. 7-5. Fetal ECG electrode placement.

low
resistance
attachment
necessary

Electrode placement techniques providing low source resistance, as covered in Chapter 16, must be used. The abdominal and back electrodes affixed to a subject are shown in Fig. 7-5. It has been reported that, during early stages of pregnancy, better results may be obtained by exerting considerable pressure onto the electrodes so as to move these electrodes closer to the actual fetus.

7.5 ELECTRICAL INTERFERENCE

away from
power lines

Particular care must be taken to eliminate line frequency interference from the fetal ECG recordings. Adequate instrumentation can eliminate any line frequency signal appearing as a common mode signal, however it is impossible to eliminate line frequency interference if it appears as a differential signal. Filters cannot be used, as the prime frequency content of the fetal ECG is between 20 Hz and 80 Hz. Line frequency interference can usually be eliminated by recording the fetal ECG with the Type 410 Physiological Monitor operated from its internal batteries and by completely disconnecting other power-line-operated devices within the near vicinity. Such devices should be completely disconnected rather than simply turned off to eliminate electrostatic interference if the switch is located in the power-line neutral lead. Interference can be eliminated in many cases by orienting the patient in a particular position within the room. If this is not successful, it is normally preferable to locate another recording site rather than to invest in costly electrostatic and electromagnetic shielding. The fetal ECG recording may still show evidence of fetal ECG activity even though line frequency interference is present. The leads connecting the patient to the monitor should be as short as possible, preferably no longer than two feet.

short leads

7.6 RECORDING TECHNIQUES

Electrodes should be placed on the mother's abdomen as shown in Fig. 7-4. If it is intended that the fetal ECG from many patients be recorded, it is desirable that some form of custom electrode selection switching unit be built to eliminate the necessity of having to make separate connections to the recording equipment for each electrode configuration. A typical set of fetal ECG's obtained by recording all combinations of the electrodes shown in Fig. 7-4 is shown in Fig. 7-6. Normally it is not necessary to record all possible combinations of these electrodes and the procedure is usually considered complete when a fetal ECG has clearly been observed in two or three configurations.

electrode combinations

In early stages of pregnancy the following electrode combinations appear to give the greatest possibility of observing the fetal ECG: C-F, B-D, A-F, A-D and, if the G and H electrodes are used, G-F, G-D and B-H. In later stages of pregnancy when the fetal ECG is pronounced, it is normally only necessary to record from either of the three sets of electrodes forming the 60° triangles of the Blondheim configuration, that is, electrodes A, F and D or electrodes B, C and E. As the amplitude of the fetal ECG is normally between 5 and 50 microvolts, and, as it is preferable to use battery-operated equipment to record the fetal ECG for reasons discussed in the previous section, most commercial electrocardiographs or physiological monitors are not compatible with fetal ECG recording. It is common practice to use an EEG recorder or a physiological monitor operated in the EEG position to monitor the fetal ECG. This technique is recommended when recording the fetal ECG with the Type 410 Physiological Monitor. The EEG position on the function selector switch is selected and the fetal ECG electrodes are connected to the EEG position on the patient cable. A standard 410, as well as any other commercially available EEG recorder, may give satisfactory results, however most of these instruments have a low frequency response extending below .1 Hz. Muscle activity within the mother's abdomen will probably result in severe interference causing the trace to be off screen most of the time. This problem can be

using the 410

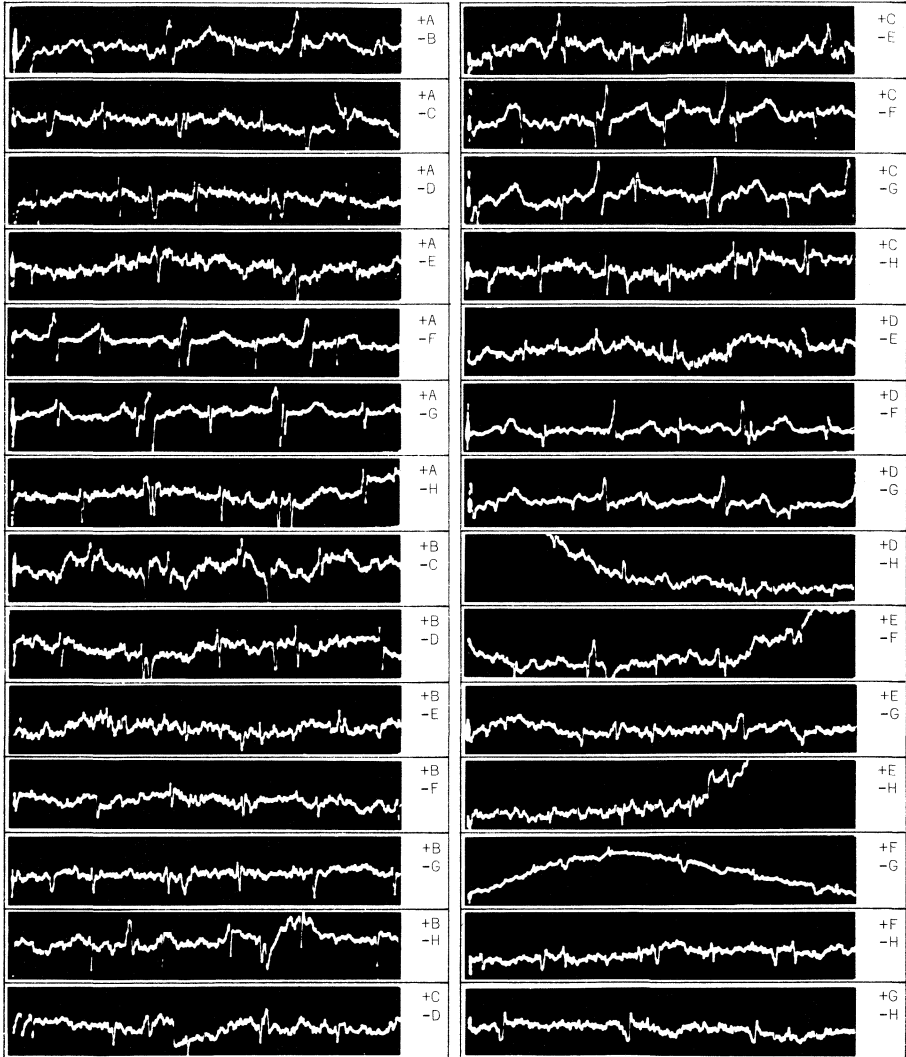


Fig. 7-6. Fetal ECG variations with various electrode positions (23 weeks gestation).

alter low-
frequency
cutoff

overcome by altering the low frequency cutoff characteristics of the monitor. A modification that may be added to the 410 monitor to increase the low frequency response in the EEG position from below .1 Hz to approximately 1 Hz is given in Chapter 29. This modification also reduces the gain in the EEG position from 50 to 100 microvolts per centimeter. The gain can, however, be increased with the variable control by a factor of 3 to give a maximum gain of 33 microvolts per centimeter. All fetal ECG photographs shown in this chapter were recorded at a sensitivity of 33 microvolts per centimeter and at a sweep speed of 50 millimeters per second.

photography

As detailed analysis of the waveform is often necessary to detect the fetal ECG, particularly in the earlier stages of pregnancy, it is desirable that a photograph be taken of the information appearing on the screen of the 410 monitor. Although no rigid mounting bezel is available to adapt the 410 monitor to a Tektronix trace-recording camera, a Tektronix C-30A camera can be held against the faceplate of the 410 monitor and satisfactory results can be achieved. All photographs shown in this chapter were recorded in this manner. When photographing from the 410 with the C-30A camera, the shutter should be in the B position and opened manually for approximately 2 seconds. Optimum results are obtained using an aperture of $f/16$ and a magnification of .9X. Certain nonlinearities exist when photographing the curved faceplate CRT used in the 410, however these nonlinearities can be compensated for by examination of the trace to identify two or more graticule lines in the photograph. Once these graticule lines are identified, the ratio of the spacing between these graticule lines to the spacing between the fetal ECG R waves gives the fetal ECG heart rate as given by the following formulas:

compensating
for nonlinear
photo

-- for fetal ECG spacings measured near the center of the screen using a sweep speed of 50 millimeters per second.

$$\text{Fetal heartrate} = \frac{300 \times \text{graticule mark spacing}}{\text{fetal ECG R wave spacing}} \times .9$$

Beats per min.

-- for fetal ECG spacings measured near the edge of the screen using a sweep speed of 50 millimeters per second.

$$\text{Fetal heartrate} = \frac{300 \times \text{graticule mark spacing}}{\text{fetal ECG R wave spacing}}$$

Beats per min.

Due to the nonlinearities mentioned earlier, a correction factor of .9 is necessary for information at center screen and no correction factor is necessary for information at the edge of the screen. Chapter 24 discusses photography of the 410 in more detail and gives the reasons for these correction factors.

7.7 INTERPRETATION OF THE FETAL ECG

When attempting to record the fetal ECG, positive results give a positive indication that the fetal ECG does exist; however negative results are nonconclusive and give no indication as to the viability of the fetus. Although the fetal ECG can be recorded as early as the 11th week and can be recorded in almost all cases by the 16th week, it is generally accepted that fetal heart activity cannot be detected by a standard stethoscope earlier than approximately the 20th week. Thus the fetal ECG can give a positive indication of the existence of a live fetus several weeks earlier than can be obtained with a stethoscope.

fetal
ECG versus
stethoscope

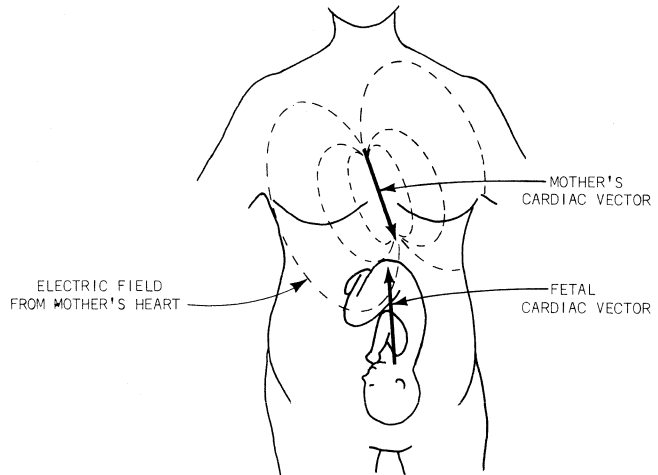


Fig. 7-7. Fetal cardiac vector vertex presentation at full term.

mother's
ECG vector

fetal ECG
vector

multiple
pregnancy

Fig. 7-7 shows the orientation of the electrical axis of the fetal heart compared to the orientation of the electrical axis of the mother's heart. As the electric field generated by the mother's cardiac vector produces "mother's ECG" activity when recording the fetal ECG, the axes of this "mother's ECG" activity will be tangential to this field and *not* necessarily parallel to the direction of the mother's cardiac vector. Within a few weeks after the fetal ECG is first detected, its amplitude is great enough for accurate measurement and the results can be plotted on a vector diagram as shown in Fig. 7-8 to determine the electrical axis of the fetal heart. Fig. 7-8 shows the electrical axes obtained from the recording shown in Fig. 7-6. As only two fetal ECG's along two axes are necessary to plot the fetal ECG vector, the other four vectors shown on Fig. 7-8 can be used to determine the accuracy of the measurement technique. An error triangle is obtained by plotting the intersection of all vectors; the head of the fetal ECG vector must lie somewhere within this error triangle.

Fetal electrocardiography is perhaps most useful when attempting to diagnose multiple pregnancy. X-ray techniques are known to be harmful to the fetus; however, fetal ECG techniques are completely harmless and can detect multiple pregnancy as early as the 16th week. Fig. 7-9 shows a fetal ECG

ALGEBRAIC SUM OF + AND - COMPONENTS OF THE FETAL ECG SHOWN IN FIG. 6 PLOTTED ON A 60° VECTOR DIAGRAM FORMED BY THE ELECTRODE POSITIONS SHOWN IN FIG. 4.

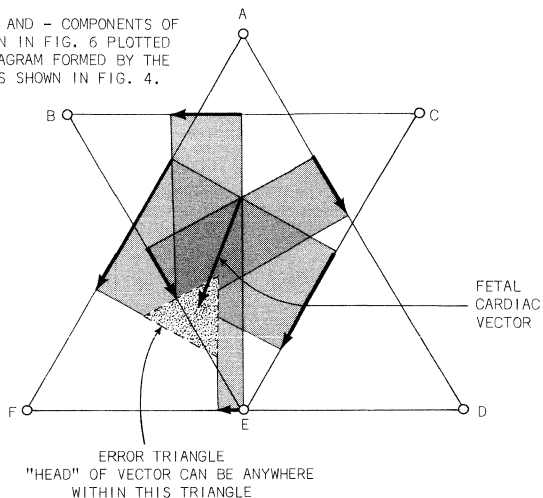


Fig. 7-8. Vector determination of fetal position.

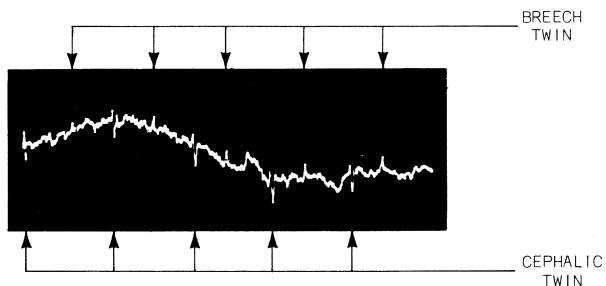


Fig. 7-9. Twin gestation — fetal ECG at 21 weeks.

recording of twins at 21 weeks. As the fetal R waves are of opposite polarity, it is obvious that their electrical vectors are in opposite directions, i.e., one twin is breech and the other is cephalic. Fetal R waves in the same direction as the mother's QRS complex represent the breech twin and fetal R waves in a direction opposite to the direction of the mother's QRS complex represent the cephalic twin.

The fetal ECG is often recorded during fetal delivery to indicate whether the delivery will be from the vertex or the breech position and to give an indication of fetal distress during labor.



BLOOD PRESSURE AND FLOW

Blood pressure measurements can be classified into three discrete groups: direct, indirect and relative. Direct blood pressure measurement involves gaining access to the circulatory system and measuring the pressure in the system directly with some form of pressure transducer. Indirect blood pressure measurement involves application of pressure external to the circulatory system and observation of the effect of this external pressure on the system. Relative blood pressure measurements are uncalibrated indirect measurements usually performed with simpler, and inherently more convenient, instrumentation. Blood flow is derived from the measurement of blood velocity which necessitates gaining access to one or more of the primary arteries within the body using surgical techniques. Information on blood pressure and blood flow is often adequate for analysis of the circulatory system, however a complete hydrodynamic analysis of the circulatory system can only be accomplished by also measuring blood volume.

8.1 DIRECT BLOOD PRESSURE MEASUREMENT

Direct blood pressure is measured by inserting a pressure transducer somewhere within the circulatory system: Figs. 2-1, 2-2 and 2-3 of Chapter 2 refer to this system. Some transducers are designed to be inserted directly into the circulatory system, however it is more common to connect to the circulatory system with either a catheter or a hypodermic needle and to record the pressure with a pressure transducer attached to the catheter or needle.

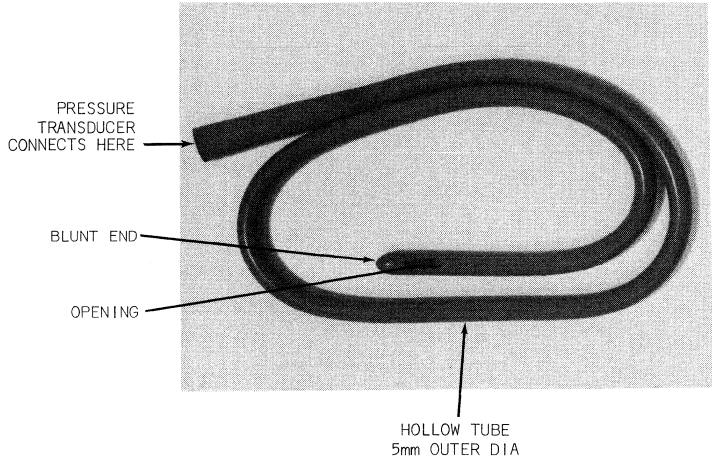


Fig. 8-1. A 5mm cardiac catheter.

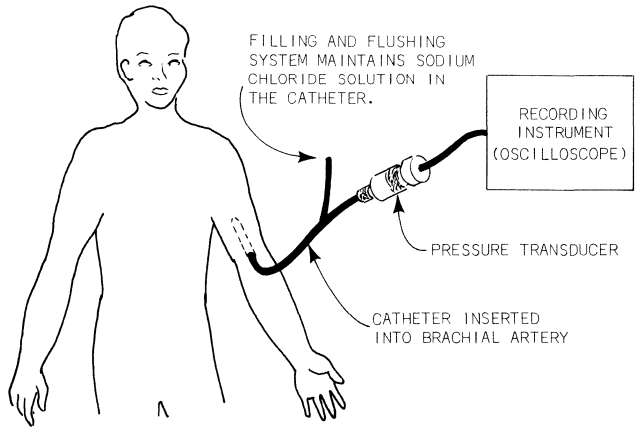


Fig. 8-2. Direct blood pressure measurement.

cardiac catheter

A cardiac catheter consists of a rubber, teflon or polyethylene tube with one end formed to a smooth bullet-like shape to allow its easy introduction into veins and arteries. A small opening is formed about 1 centimeter from this smooth end as shown in Fig. 8-1. The catheter shown in Fig. 8-1 is approximately 40 centimeters long and 5 millimeters in diameter. Catheters are generally available from 1 to 10 millimeters in diameter and in various lengths. Cardiac catheters are also often described by their circumference.

entry to
circulatory
system

With the catheter/pressure-transducer configuration shown in Fig. 8-2, pressures can be measured in almost any portion of the circulatory system. By gaining entry to the circulatory system at either the arms or legs a catheter may be manipulated throughout the circulatory system. It should be noted that the pressure measured will be the pressure at the blunt end of the catheter and not the pressure at the point where the catheter enters the circulatory system.

cardiac
catheteri-
zation

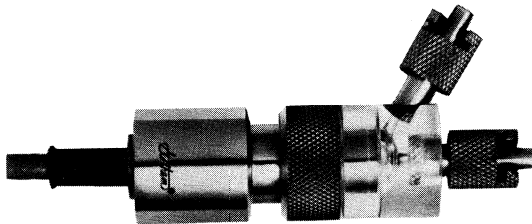
It is possible, with careful manipulation, to actually introduce the smooth end of the catheter into the right atrium via the inferior vena cava. Cardiac defects, including holes in the heart, can then be explored. This procedure is referred to as cardiac catheterization. Investigation of the left ventricle by catheterization via the aorta, referred to as arterial catheterization, almost always causes trouble with arterial damage and leakage into the tissues.

arterial
pressure

The most common direct pressure measurement is arterial pressure which is usually measured by introducing the catheter into the brachial artery at the elbow of either arm.

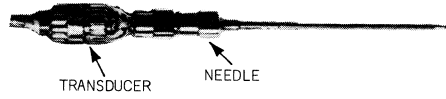
transducer

A typical blood pressure transducer for use with a catheter is shown in Fig. 8-3. As little difference in pressure is encountered throughout the arterial system, it is unnecessary to introduce the catheter any great distance into this circulatory system.

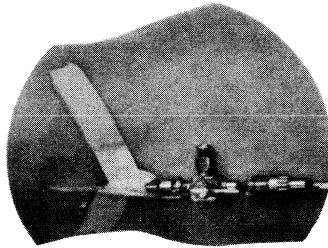


STATHAM INSTRUMENTS INC. - MODEL P23De
LENGTH - 2.5"
OUTPUT WITH 7.5VRMS EXCITATION = 75 μ VRMS/mmHg
FOR USE WITH CATHETERS.

Fig. 8-3. A conventional blood pressure transducer.

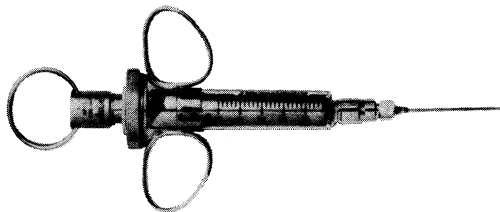


STATHAM INSTRUMENTS INC. - MODEL P37
 TRANSDUCER LENGTH \approx 1.2"
 OUTPUT WITH 7.5VRMS EXCITATION = 37μ VRMS/mmHg



TRANSDUCER CONNECTED AT BRACHIAL
 ARTERY PUNCTURE SITE.

Fig. 8-4. A needle blood pressure transducer.



STATHAM INSTRUMENTS INC. - MODEL SP25
 OVERALL LENGTH \approx 9"
 OUTPUT WITH 7.5VRMS EXCITATION = 37μ VRMS/mmHg

Fig. 8-5. A syringe blood pressure transducer.

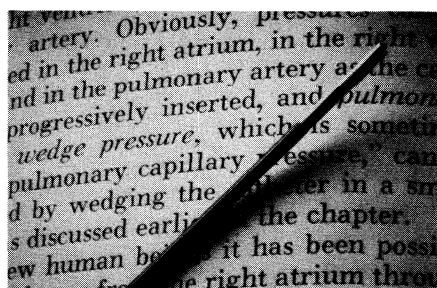
needle/
transducer

The catheter/transducer combination may be replaced by a needle and transducer as shown in Fig. 8-4. The needle/transducer combination is also available in a syringe configuration as shown in Fig. 8-5. These combinations, using needles rather than flexible catheters, are obviously far easier to introduce into the subject's circulatory system and are thus preferable for direct blood pressure measurement. They are normally introduced into the femoral artery near the groin as this artery is close to the surface at this point. The syringe configuration allows administration of an arterial injection while recording arterial pressure.

drift
catheters

For measurement of blood pressure within the heart, very small diameter (1 millimeter) catheters, known as drift catheters, are required so as not to interfere with the normal operation of the heart. These small catheters are inserted into the venous system and "drift" with the blood flow, eventually reaching the heart. They may actually be passed through the heart into the pulmonary circulatory system. The primary disadvantage of these small catheters is the damping effect they have on the pressure pulse waveform which limits the high frequency response of the system and gives erroneous readings. To avoid this problem, the catheter tip transducer shown in Fig. 8-6 incorporates a small pressure transducer built into the end of the catheter, the combination being less than 2 millimeter in diameter. This allows introduction of the actual transducer into the heart and thus avoids the damping problem inherent with small catheters.

catheter tip
transducer



STATHAM INSTRUMENTS INC. - MODEL P866
 TRANSDUCER TIP - .065" DIAMETER
 OUTPUT WITH 7.5VRMS EXCITATION = 5 μ VRMS/mmHg

Fig. 8-6. A catheter tip pressure transducer.

bridge-type
transducers

Recording from transducers is covered in more detail in Chapter 18. The transducers shown in this chapter are all bridge-type transducers and, when excited at their maximum rated excitation voltage, they produce an output in the low millivolt or microvolt region, the smaller transducers in general producing less output. While it is possible to use DC excitation and record the output directly using a high sensitivity amplifier, superior results are obtained using a carrier amplifier system such as the Tektronix 3C66 plug-in.

micro-
electrode
pressure
transducer

Extremely small direct blood pressure "transducers" have recently been developed for research work. These transducers have an effective diameter of 10 microns. The transducer consists of a 10 micron microelectrode filled with an electrolyte having a different resistivity than that of blood or body fluids. A pressure system is connected to the microelectrode and the impedance between the center of this electrode and the body is monitored. A servo control system controls the pressure in the microelectrode driving pressure system to keep the impedance between the electrode and the body constant by exerting a pressure on the electrolyte in the microelectrode equal to the pressure external to the microelectrode tip. This pressure exerted on the electrolyte is therefore a measure of the blood pressure at the tip of the microelectrode.

measurement
terminology

Arterial pressures are normally measured as millimeters of mercury, mm/Hg, and venous pressures are usually measured as centimeters of water, cm/H₂O. These units are derived from older blood pressure measuring techniques using either mercury or water manometers. A typical arterial blood pressure waveform is shown in Fig. 8-7. The peak pressure is referred to as the systolic pressure and the minimum pressure is referred to as the diastolic pressure.

It is common terminology to refer to an arterial blood pressure of, say, 120 millimeter systolic and 80 millimeter diastolic as "120 over 80" or 120/80. As stated in Chapter 2, the pressure in the arteries is many times greater than the pressure in the veins, typical arterial pressure being 120/80 and typical venous pressure being 9/5 mm/Hg or, in the more common terminology, 12/7 cm/H₂O.

While direct blood pressure is relatively easy to measure, it does require an incision in an artery for introduction of a catheter or insertion of a hypodermic needle. Since this often results in the permanent loss of the artery used, it is regarded with some disfavor in most countries except the U.S.A. Results accurate enough for clinical use are usually obtainable by indirect methods.

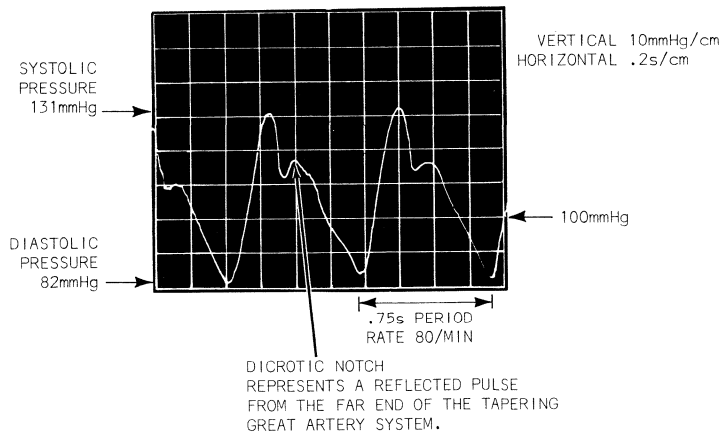


Fig. 8-7. Arterial blood pressure waveform.

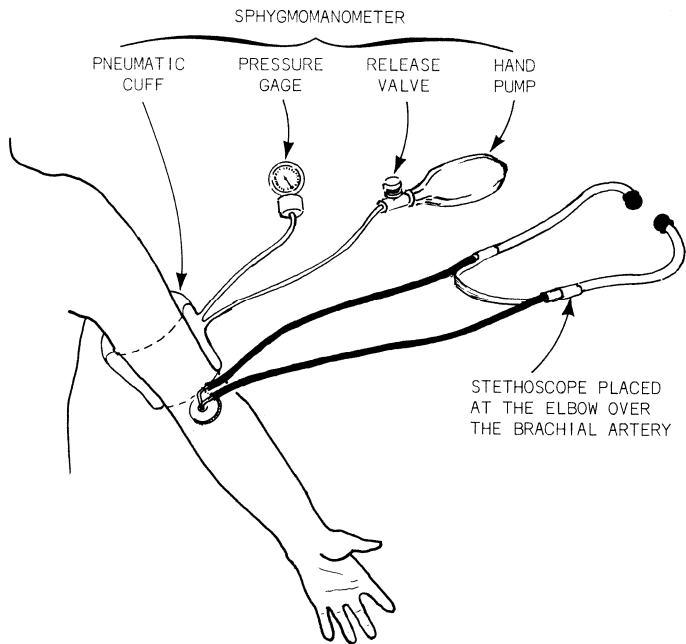


Fig. 8-8. Indirect blood pressure measurement with a sphygmomanometer.

8.2 INDIRECT BLOOD PRESSURE MEASUREMENT

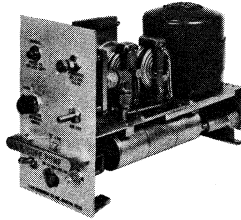
sphygmoma-
nometer

By far the most common form of blood pressure measurement is the indirect measurement using the familiar pressure cuff, hand pump and pressure dial device, used by all physicians, referred to as a sphygmomanometer. The sphygmomanometer, as shown in Fig. 8-8, incorporates a pneumatic cuff encircling the upper arm. An inflatable section of this cuff is inflated by a small hand pump and the pressure in the system is indicated by a mechanical pressure gauge or, in some models, a mercury manometer. The cuff is inflated to a pressure greater than the blood pressure in the large brachial artery of the arm. This pressure thus collapses the artery and occludes (cuts off) blood flow to the arm. As the pressure in the cuff is gradually released using a release valve built into the hand pump, a point is reached where the cuff pressure and the peak or systolic arterial pressure are the same. At a pressure slightly below this level the peak arterial pressure slightly exceeds the cuff pressure and blood is able to squirt through the compressed segment of the brachial artery. This squirting blood results in turbulence within the artery creating sounds known as "Korotkoff" sounds. These sounds are usually detected with a stethoscope placed over the brachial artery. As the pressure in the cuff is further decreased, Korotkoff sounds continue until a point is reached where no further turbulence is produced as no constriction exists in the brachial artery. This point represents the diastolic blood pressure. As it is somewhat difficult to detect the pressure where the Korotkoff sounds begin and cease, this sphygmomanometer technique cannot be relied upon to produce an accuracy of much better than about 10 millimeters of mercury. While the technique is inaccurate, it is simple to perform and very little discomfort is felt by the patient. In the hands of a skilled operator highly repeatable results are obtained and, since the clinician is usually more interested in trends than exact numbers, the technique is entirely appropriate.

Korotkoff
sounds

PUMP MAY BE OPERATED FOR A SINGLE CYCLE OR MAY BE PROGRAMMED TO CYCLE AT PRESET INTERVALS.

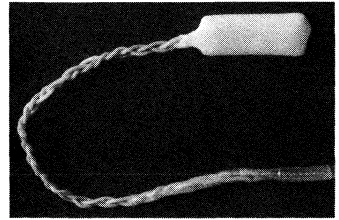
A CONTACT CLOSURE DURING INFLATION CAN BE USED TO ERASE A STORAGE OSCILLOSCOPE.



E & M INSTRUMENT CO., INC. 95-300-70

Fig. 8-9. An automatic cuff pump.

PIEZOELECTRIC CRYSTAL PROVIDES HIGH OUTPUT LEVELS



E & M INSTRUMENT CO., INC. 92-201-70

Fig. 8-10. A Korotkoff sounds microphone.

automatic
measuring
equipment

The sphygmomanometer technique may be automated by replacing the hand pump with an automatic cuff pump as shown in Fig. 8-9. The automatic cuff pump may be operated by pushing a panel-mounted button to produce a single cycle of inflation and deflation or it may be set for repeat cycles at various intervals for continuous monitoring of blood pressure over long periods of time. The stethoscope may be replaced by a Korotkoff-sound microphone as shown in Fig. 8-10. This microphone consists of a small piezoelectric transducer specifically designed to efficiently reproduce Korotkoff sounds. The pressure indicating dial may also be replaced with a pressure transducer similar to the transducers used for direct pressure measurement.

using an
oscilloscope

The subject's indirect blood pressure may be conveniently recorded using the system shown in Fig. 8-11. The vertical channel of an oscilloscope displays the output from a Korotkoff sound microphone while the horizontal channel displays the output from a pressure transducer. A typical display produced by this system is also shown in Fig. 8-11. The horizontal axis is calibrated in pressure; the points at which vertical information appears and then disappears are the systolic and diastolic pressures. This system is capable of somewhat greater accuracy than the conventional cuff/stethoscope as it removes human judgment in determining the presence of Korotkoff sounds. In some cases clearer displays have been obtained by rejecting all information from the microphone below 150 Hz using a high-pass filter between the microphone and the oscilloscope vertical channel. The prime frequency content of this information appears to be between 400 Hz and 500 Hz.

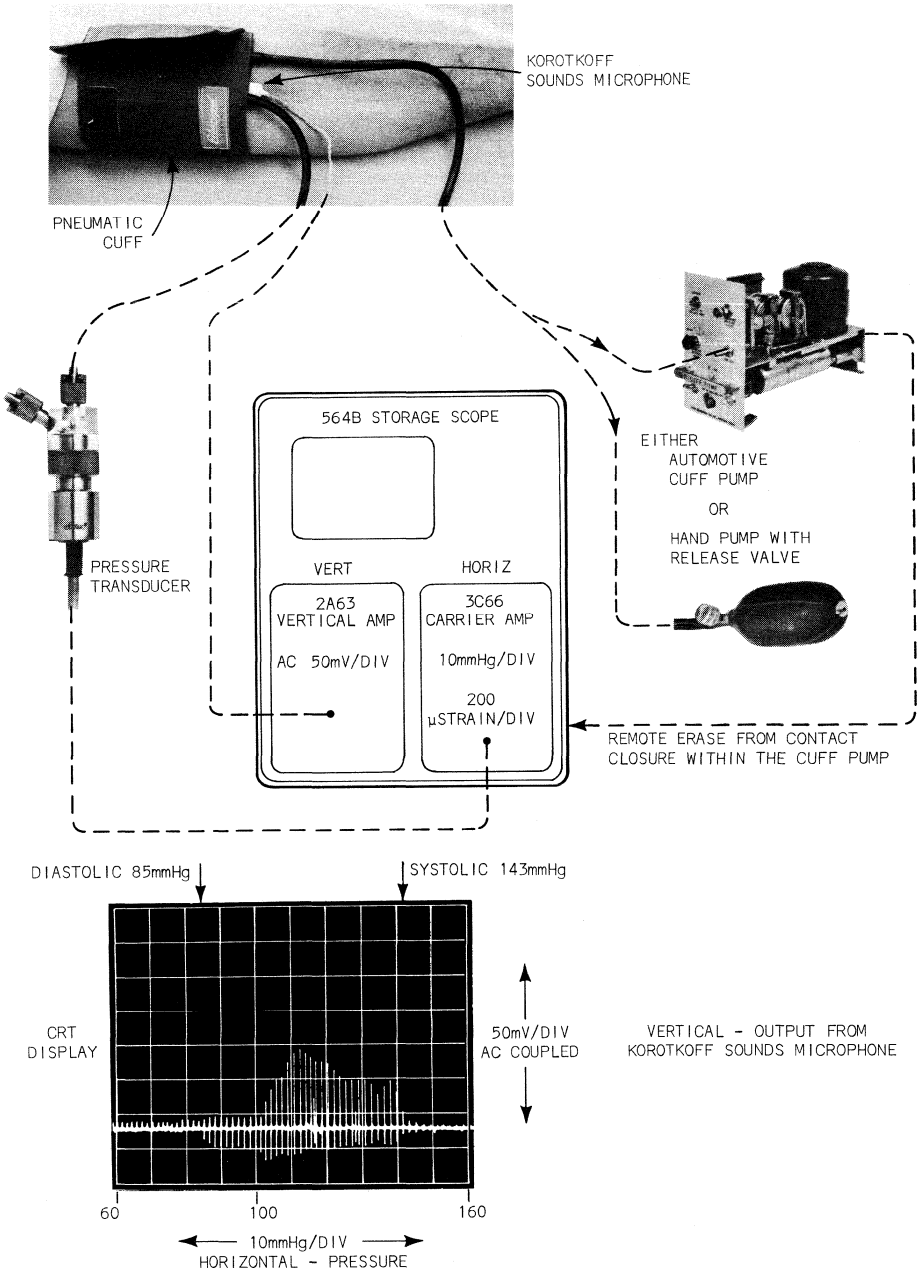
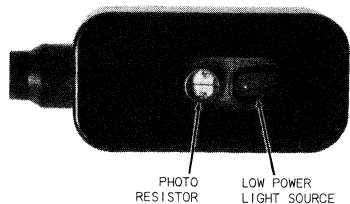


Fig. 8-11. Automatic indirect blood pressure measurement.

8.3 INDIRECT RELATIVE BLOOD PRESSURE MEASUREMENT

In many cases, it is unnecessary to measure the absolute value of arterial blood pressure; all that is required is an indication of blood flow throughout the body. If blood flow throughout the body diminishes for some reason, the principal areas of the body deprived will be the fingers and the toes, thus monitoring the presence of blood flow in these areas will insure that blood is indeed flowing throughout the body. The simplest technique of recording the presence of blood flow in the peripheral arteries is to use a plethysmograph. Plethysmography is the art of monitoring the physical changes in size of part of the body as modified by the flow of blood within it. Various techniques are used to detect this change in size. The Pulse Sensor utilizes a photoelectric technique. Other techniques, such as impedance measuring, are also used to indicate relative change in size.

The Pulse Sensor uses a light source and photodetector to record a change in opacity of the flesh as blood is pumped through it. This Pulse Sensor relies on light being reflected by some reflecting medium such as bone, etc. For adequate output potentials to be obtained, a fairly high concentration of arteries near the surface is required. The Pulse Sensor produced by Tektronix for use with the Type 410 Physiological Monitor is ideally suited for application to the finger and is less suitable for toe, ear or nose use. If the subject is in a drugged state, or if he is particularly cold, then vasoconstriction will occur which tends to limit blood flow to the limbs. Under these conditions it may be preferable to tape the Pulse Sensor to the forehead of the subject. It is imperative that the Tektronix Pulse Sensor be used with resting subjects and, as it detects changes in physical size, it is definitely not suited to exercising subjects. The Tektronix Pulse Sensor is shown in Fig. 8-12, and is applied to the finger and forehead as shown in Fig. 8-13. A typical plethysmogram recorded with this Pulse Sensor and the Tektronix Type 410 Physiological Monitor is shown in Fig. 8-14.



RESISTOR AND LIGHT SOURCE ARE SEEN THROUGH "WINDOW" IN THE PULSE SENSOR

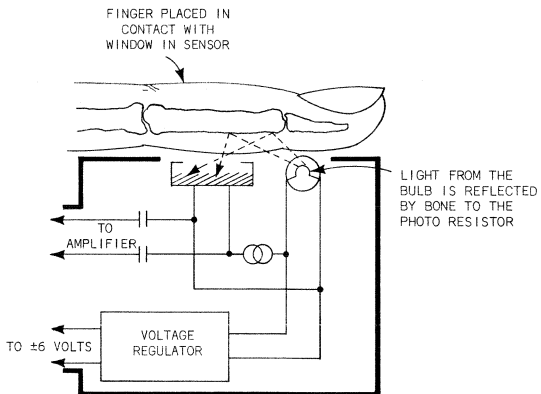
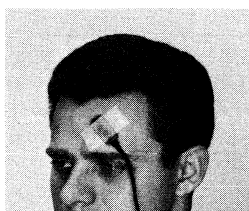
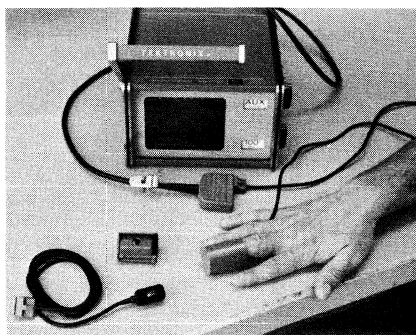


Fig. 8-12. The Tektronix plethysmograph (pulse sensor).



PLETHYSMOGRAPH HELD IN PLACE BY TAPE OR BY AN ELASTIC STRAP AROUND THE HEAD.



PLETHYSMOGRAPH HELD IN PLACE BY THE TEKTRONIX FINGER HOLDER.

Fig. 8-13. Placement of the Tektronix plethysmograph.

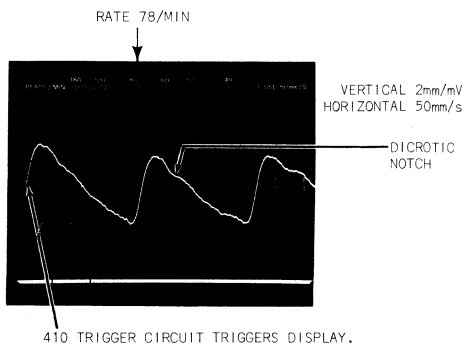


Fig. 8-14. Plethysmogram obtained with Tektronix Pulse Sensor (attached to finger) and 410 Monitor.

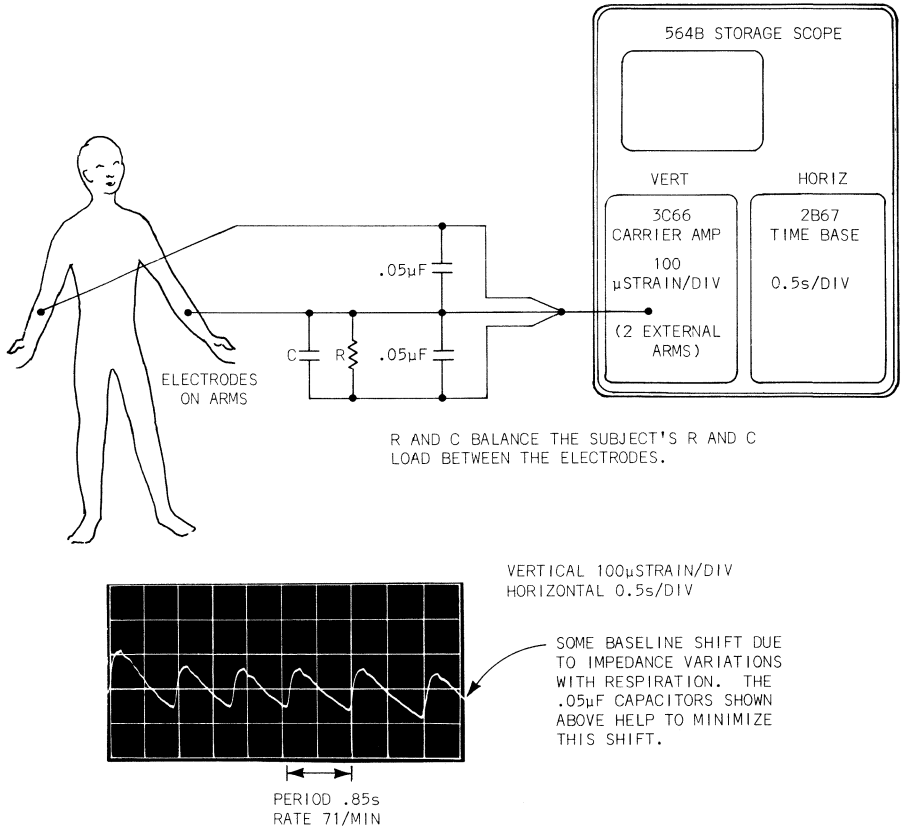


Fig. 8-15. Impedance plethysmography with Tektronix 3C66 Plug-In.

impedance
measuring

As stated above, impedance measuring techniques may also be used to indicate relative changes in size resulting from blood flow. If the impedance between the arms is monitored, it will be found that the impedance changes due to the action of the heart and due to respiration. Extensive studies have shown that the sensation level of arm to arm electrical current increases with frequency: some sensation may be felt with 1 mA of DC current however, at about 25 kHz, ten times this current is required to produce sensation. Thus audio frequency currents of at least an order of magnitude less than 1 mA are used to monitor impedance. These frequencies and low currents also avoid cell stimulation. It has been found that when measuring the impedance between the arms, cardiac action primarily changes the resistive component of this impedance and respiratory action primarily changes the capacitive components. Thus, to detect cardiac action, one must use an audio frequency system sensitive only to resistive changes, to avoid respiratory effects.

Such a system can be incorporated using the Tektronix Type 3C66 Carrier Amplifier with a storage oscilloscope and time base as shown in Fig. 8-15. With this system capacitive impedance changes are almost completely eliminated by paralleling the impedance to be measured with a large fixed capacitor. A similar capacitor is required to balance the bridge, as is a balancing R and C to simulate the R and C of the subject. The 3C66 is operated in a Wheatstone Bridge configuration with two of the arms of the Wheatstone Bridge external to the instrument, the other two internal. A typical impedance plethysmograph recorded with the above technique is also shown in Fig. 8-15. The slight change in base line evident in this recording is due to the influence of respiration. Further details on impedance measuring techniques are given in Chapters 9, 13 and 18.

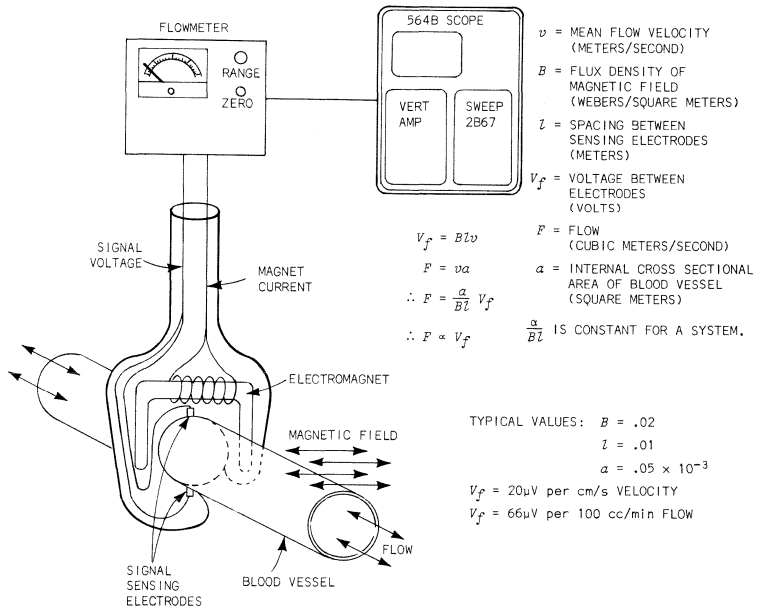


Fig. 8-16. Electromagnetic flowmeter – theory.

8.4 BLOOD FLOW MEASUREMENT

electromagnetic velocity transducer

Blood flow is measured by placing a mean velocity transducer in an artery having a known cross-sectional area. Blood flow is the product of the mean velocity and area. Many types of mean velocity-sensitive blood flowmeters have been developed; the electromagnetic blood flowmeter is now extensively used. A diagrammatic representation of an electromagnetic blood flowmeter is shown in Fig. 8-16. The theory of electromagnetic flowmeters is based on Faraday's law. When a conductive fluid, such as blood, traverses the lines of force of a magnetic field, an electromotive force is generated in the fluid which is perpendicular to both the magnetic lines of force and the direction of motion of the fluid. This electromotive force is directly proportional to the intensity of the magnetic field, the distance between the sensing electrodes and the fluid velocity.

The electromagnetic blood flow transducer consists of an electromagnet to generate a magnetic field and two electrodes to sense the flow signal. They are encapsulated in epoxy in a form to allow them to fit around the blood vessel. The lumen or inside diameter fixes the cross-sectional area of the vessel, changing the transducer to a flow-rate measuring instrument although basically it is a velocity transducer. Electrodes make contact with the vessel wall. The flow transducer is connected to a flowmeter. This flowmeter supplies energizing current to the electromagnet, amplifies the flow signal, discriminates it from artifacts and makes it available for display on an oscilloscope.

flow-rate
measurements

flowmeter

The above theory assumes a DC magnetic field which would produce a DC flow signal. Since it would be impossible to differentiate this signal from electrode offset potentials, amplifier drift, etc., commercial blood flowmeters use AC excitation; either sinewave or squarewave. When revising the above theory for AC signals a $\frac{\delta B}{\delta t}$ component appears in the output voltage formula. Commercial systems measure the output voltage only during the period when this $\frac{\delta B}{\delta t}$ component is equal to zero. In sinewave excitation, the $\frac{\delta B}{\delta t}$ component is effectively zero only for a relatively short time at the peak of the excitation wave. The voltage sensing circuit must gate on during this time and correct adjustment of the gate is critical. In the squarewave excitation, the $\frac{\delta B}{\delta t}$ component is always zero except during short switching periods, thus larger variations in gate characteristics are permissible.

AC
excitation

sinewave

squarewave

signal-to-
noise ratio

power
requirements

frequencies

Signal-to-noise ratio is proportional to the peak-to-peak value of the excitation voltage. For sinewave and squarewave excitation of the same peak-to-peak voltage, the power required for sinewave excitation is only 50% of the power required for squarewave excitation. The chief disadvantage of squarewave probes is, therefore, that they are larger and must operate at higher temperatures than sinewave probes to maintain an acceptable signal-to-noise ratio. Both systems require excitation frequencies high enough to permit effective sampling, typically from 200 Hz to 1000 Hz.

commercial
flow probes

Some commercial electromagnetic blood flowmeters are shown in Fig. 8-17. The flowmeters are of the intracorporeal type which have a slot opening to allow placement around the blood vessel. These openings are usually fitted with a cover to maintain a constant blood vessel size. Extracorporeal transducers have tubular sleeve or cannula extensions and are applied by cutting the blood vessel and inserting the tubular sleeve in series with this vessel. Blood is in direct contact with the sleeve lumen in extracorporeal units.

commercial
flowmeters

Blood flowmeters suitable for use with the above transducers are produced by several manufacturers. In general, these instruments provide either sinewave or squarewave excitation at between .1 and 1 amp and provide an output for use with an oscilloscope or other recording device of between .1 volt and 1 volt. This recording device should have a response from DC to 100 Hz. The instruments are calibrated to read in cubic centimeters per minute and usually cover a range from 1 to 100,000 cubic centimeters per minute. Blood velocity in the pulmonary artery may reach 100 centimeters per second. As this artery has a cross-sectional area of about 1.8 square centimeters, this velocity results in a peak flow of 100,000 cubic centimeters per minute. Blood flow systems may be calibrated by allowing blood to escape into a graduated flask for a known brief period or by using an external calibration system providing a known flow. A zero flow baseline may be obtained by occluding flow for a brief period. A typical blood flow waveform is shown in Fig. 8-18, and a complete measuring system is shown in Fig. 8-19.

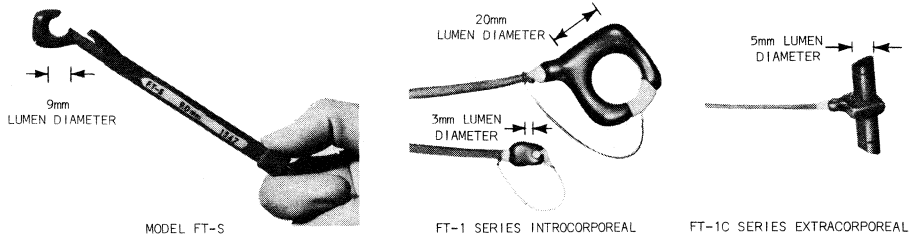


Fig. 8-17. Commercial blood flow meters. (In Vivo Metric Systems, Los Angeles, Calif.)

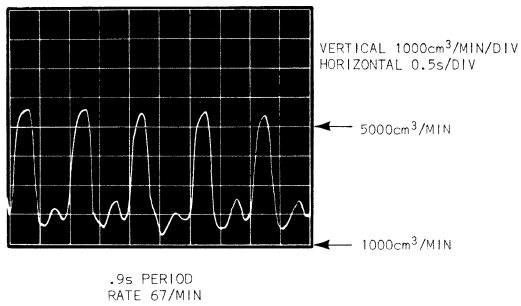


Fig. 8-18. A typical blood flow record from an electromagnetic flowmeter.

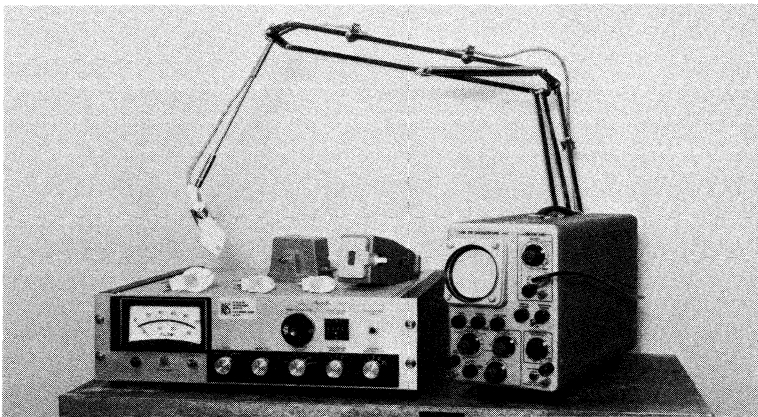


Fig. 8-19. A commercial blood flow measuring system.

other flow-measuring devices Other forms of blood flow-measuring devices are occasionally used. The isothermal flowmeter places a thermistor within the blood flow. The flowing blood tends to cool the thermistor and a measurement of blood flows is obtained by recording the increase in thermistor excitation required to maintain the thermistor at a constant temperature (constant resistance). The electro-turbine meter inserts a minute rotational generator into the blood flow system; this generator is driven by a small propeller driven by the blood flow. Ultrasonic blood flow measuring techniques are becoming increasingly popular; these techniques involve the detecting of small phase differences between ultrasonic signals propagated in the direction of flow and in the opposite direction.

8.5 CARDIAC OUTPUT

direct determination Blood flow may be measured in almost any part of the circulatory system by using the blood flow measurement techniques discussed previously. When blood flow is measured in the pulmonary artery or the aorta, this blood flow, integrated over a cardiac cycle, represents the total amount of blood flowing through the heart and is referred to as cardiac output. Thus cardiac output may be determined by integrating the results obtained from conventional blood flow measurement at the pulmonary artery or aorta. As this obviously involves critical surgical procedures, cardiac output is more commonly determined using dilution techniques as discussed briefly below. These dilution techniques give total cardiac output whereas flow measurement shows the discrete changes in cardiac output over one complete cardiac cycle.

dilution techniques

The Stewart-Hamilton dye dilution technique involves introduction of a known volume of dye into the superior vena cava via vessels in the arms and withdrawal of samples of the blood/dye mixture from the aorta via a catheter. The dilution of dye contained in these samples is determined with a densitometer. Modern dye dilution techniques use radioisotopes instead of dye and counters instead of densitometers. The output from the densitometer or counter is coupled to an analog computer which computes cardiac output directly. Cardiac output in an adult male is approximately 5000 cubic centimeters per minute.

8.6 BLOOD VOLUME

Blood volume is measured using a modified dye dilution technique. A known amount of dye is introduced into the system and allowed to circulate for many cycles. After several minutes, a blood sample is taken and the dilution of the dye in this sample is noted. Total blood volume is then a product of the dilution ratio and the original quantity of dye injected. Blood volume in an adult male is approximately 5500 cubic centimeters or one and one-half U.S. gallons.

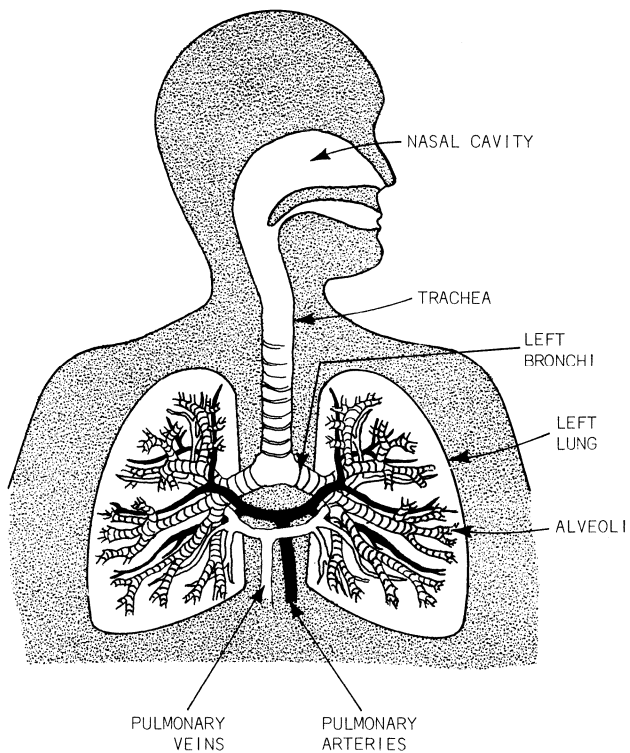


Fig. 9-1. Respiratory system.

9

RESPIRATION AND TEMPERATURE

9.1 PHYSIOLOGICAL CONSIDERATIONS

respiratory functions

The physiology of the respiratory system is not covered in Section I; thus the more important physiological aspects of this system are discussed below. The primary functions of the respiratory system are to oxygenate the blood, i.e., to dissolve oxygen into the blood, and to remove carbon dioxide from the blood. If blood is not oxygenated due to failure of the circulatory system, the oxygen content of the blood will rapidly decrease and after about 60 to 90 seconds the subject will become unconscious, death occurring in 4 to 5 minutes.

lung structure

The lungs are the major component of the respiratory system and it is in the lungs that oxygenation of the blood occurs. When the lungs are forced to expand by muscular contraction of the diaphragm and expansion of the thoracic cage by contraction of the rib muscles, air enters the lungs via the bronchi and is diverted to millions of small air sacs known as alveoli. The membrane comprising the alveoli is moist and the oxygen contained in the air is dissolved by this moisture. Interspersed with the alveoli are fine capillaries, branching from the circulatory system, through which blood is continually flowing. The oxygen dissolved in the moist surface of the alveoli diffuses into the blood stream via these capillaries. The carbon dioxide contained in the blood stream is also diffused through the alveolar membrane to be expelled with expired air. Some of the principal components of the respiratory system are shown in Fig. 9-1.

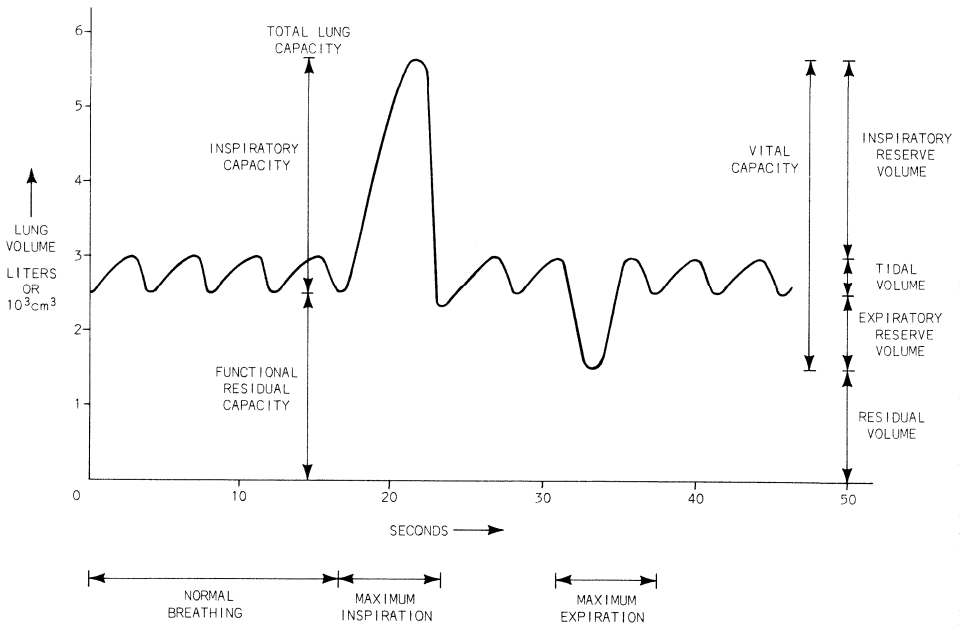


Fig. 9-2. Respiratory changes in lung volume.

lung capacity A grown man at rest inhales about 500 cubic centimeters of air with each breath and takes between 12 and 18 breaths per minute. During periods of moderate exercise he may inhale one liter or more with each breath and take up to 25 breaths per minute. At maximum exercise the vital capacity of 4 to 5 liters is reached.

lung volume The action of breathing is controlled by muscular action causing the volume of the lung to increase and decrease. During normal breathing the lung does not contract to its minimum possible volume, nor does it expand to its maximum possible volume. Minimum lung volume is obtained when exhaling with extreme effort to expel the maximum possible amount of air and maximum lung volume is obtained when inhaling with maximum effort. Fig. 9-2 shows diagrammatically the changing lung volume that may be expected for a resting man and the lung volume that may be achieved during maximum inspiratory and expiratory effort.

spirometer Lung volume is measured by a spirometer and the recording of lung volume changes with time is known as a spirogram. Instruments that simply detect respiratory activity are referred to as pneumographs and the resulting recording of respiratory activity changes with time is known as a pneumogram. A spirogram is normally only required when attempting to analyze the respiratory system or to detect a malfunction of this system and is rarely used for routine monitoring. A pneumogram may be used for routine monitoring and is basically used to indicate the fact that the subject is breathing. Breathing rate may be obtained from either a spirogram or a pneumogram.

pneumograph

9.2 RESPIRATORY ACTIVITY

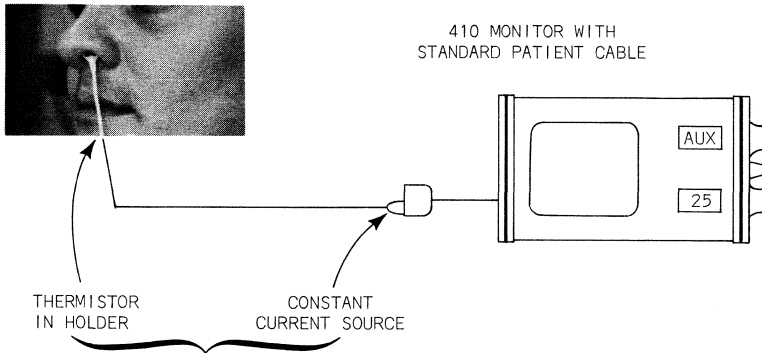
Relative respiratory activity may be detected by either detecting the physical changes in the torso associated with breathing or by detecting the flow of air through the nostrils. Since no absolute measurements are required, the measurement techniques involved have been simplified to allow easy application of the devices concerned to the subject.

Perhaps the simplest form of pneumograph employs a thermistor placed in the outer nasal passage to detect the temperature differential between inspired cool air and expired warm air. This technique fulfills the majority of clinical needs including those of operative and post-operative subjects. If the subject breaths through his mouth, or if he wishes to converse, the thermistor may be placed in the mouth or in such a position as to detect flow from either the nose or the mouth. The thermistor concerned should be supplied from a constant current source at a low enough current to maintain thermistor self-heating below one degree centigrade or so. Adequate sensitivity can usually be obtained with 5 milliwatts of thermistor dissipation. Excessive thermistor heating can cause subject discomfort, thus thermistor dissipation should be limited to 40 milliwatts for small bead-type thermistors. A thermistor pneumograph suitable for use with a Tektronix 410 monitor, together with a typical pneumogram obtained from this system, is shown in Fig. 9-3. Further details are given in Chapter 29.

thermistor
pneumograph

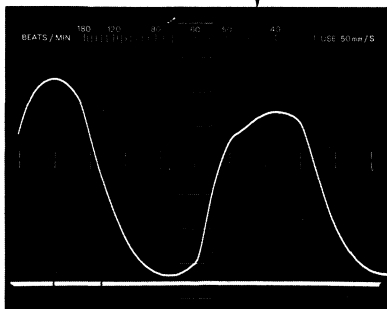
temperature
consider-
ations

If the temperature of the outside air is the same as the temperature of the expired air (body temperature) the above system is unsatisfactory. In this case, enough current should be passed through the thermistor concerned to raise its temperature in still air to a temperature somewhat above body temperature but below a temperature that may cause subject discomfort. This can usually be achieved with thermistor dissipations of between 5 and 25 milliwatts. The flow of both inspired and expired air over this thermistor will tend to cool it, thus producing a resistance change. Since both inspiration and expiration decreases the thermistor temperature, the resulting output will be at twice the respiratory frequency.



THERMISTOR PNEUMOGRAPH (FOR DETAILS SEE CHAPTER 29)

$$\text{RATE } \frac{44}{2} = 22 \text{ BREATHS/MIN}$$



VERTICAL AUX 2mm/mV = 5mV/DIV
HORIZONTAL 25 mm/s = .4s/DIV

Fig. 9-3. Thermistor pneumogram with the Tektronix 410 Monitor.

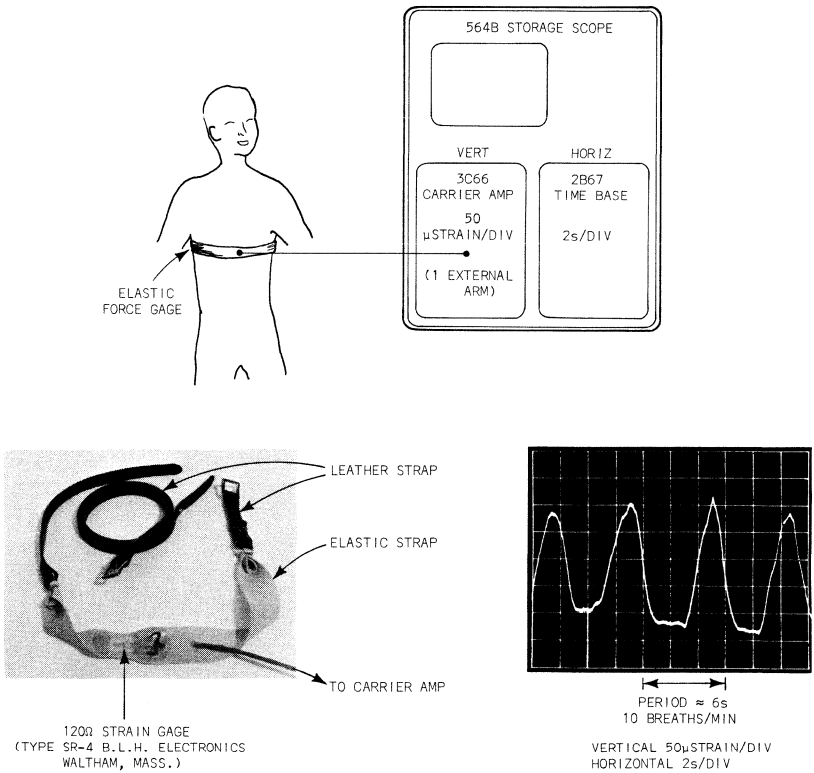


Fig. 9-4. Elastic force gage pneumogram with a Tektronix 3C66 Plug-In.

strain gauge detects chest size changes

Changes in the physical size of the torso with respiration may also be detected to indicate respiratory activity. A strain gauge attached to a piece of elastic may be used to form a band around the chest. Since chest circumference varies with the respiration, the elastic band will stretch causing a change in resistance in the strain gauge. This resistance change may be detected by a DC-excited Wheatstone Bridge or, preferably, by a carrier amplifier such as the Tektronix 3C66 system. Such a system and the resulting pneumogram obtained is shown in Fig. 9-4.

resistance changes with chest circumference

Changes in chest circumference may also be detected by a rubber tube filled with mercury, fastened firmly around the chest. As the chest expands, the rubber tube increases in length and thus the resistance of the mercury from one end of this tube to the other changes. This resistance change may be detected using a constant-current source in the same way as the thermistor resistance change was detected in Fig. 9-3. The principal disadvantage of this mercury-filled rubber tube pneumograph is its extremely low resistance, requiring large currents and sensitive detecting instruments. These disadvantages can be largely overcome by replacing the mercury with a conductive solution, such as copper sulphate solution (with copper plugs in the ends of the rubber tube) or with some of the less-viscous types of electrode paste commonly used to apply electrodes to the skin. Commercial electrode pastes vary greatly in resistivity; however, a 45-inch long 1/8 inch diameter rubber tube filled with electrode paste should have a resistance of between 1,000 and 100,000 ohms.

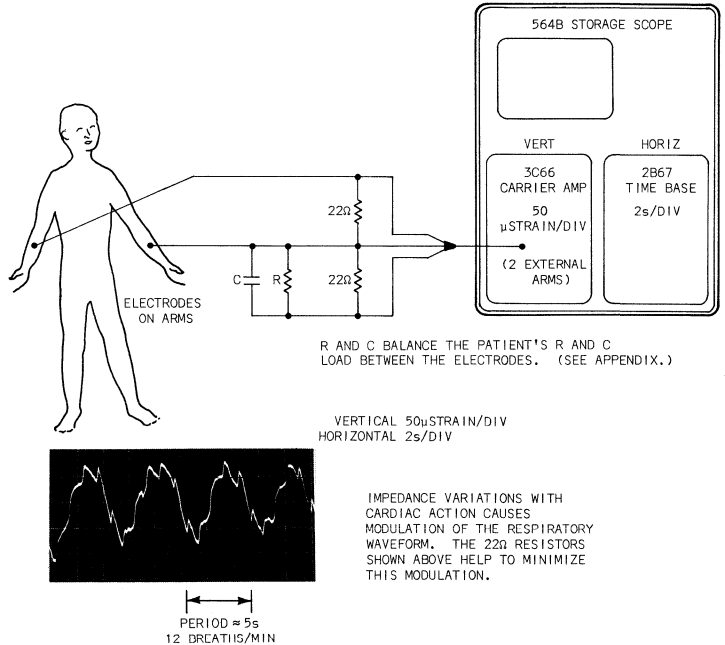


Fig. 9-5. Impedance pneumogram with a Tektronix 3C66 Plug-in.

torso resistivity

Respiratory activity may also be detected by measuring the change in resistivity across the torso. This technique is similar to the technique used to measure cardiac activity as described in Chapter 8, Section 8.14. However, whereas the resistive components of this impedance are detected to measure cardiac activity, the capacitive component of this impedance is detected to indicate respiratory activity. A typical impedance pneumograph system together with a typical impedance pneumogram obtained from this system is shown in Fig. 9-5. There appears to be little practical use for this system as the variations produced by the heart in the record prevent a good recording from being obtained.

9.3 RESPIRATORY AIR FLOW

pneumotach

Respiratory flow is invariably measured with a pneumotachograph (commonly referred to as a pneumotach), consisting of a hydraulic resistance head, and a differential pressure transducer. The pneumotach consists of a one-inch diameter tube containing a fine mesh screen, as shown in Fig. 9-6.

This mesh screen offers slight resistance to air flow; this resistance to flow produces a pressure differential across the mesh screen which is proportional to the mean flow velocity. This pressure differential is detected by a differential pressure transducer. The sensitivity of the device can be varied by varying the size of the mesh screen or by using several mesh screens, however total airway resistance offered to a subject should never exceed about 1 cmH₂O if normal respiration is not to be affected by the measuring device. A typical pressure differential across the screen would be 0.09 cmH₂O per 10 liters per minute flow using 2 inch diameter, 400 per inch, stainless steel gauze in the head.

measurement
system
calibration

The differential pressure transducer can be used in conjunction with a Tektronix Type 3C66 Carrier Amplifier in a Tektronix Type 564B Storage Oscilloscope. The complete respiratory flow measuring system, i.e., pneumotach, differential pressure transducer, 3C66 carrier amplifier and oscilloscope, may be calibrated with a known flow.

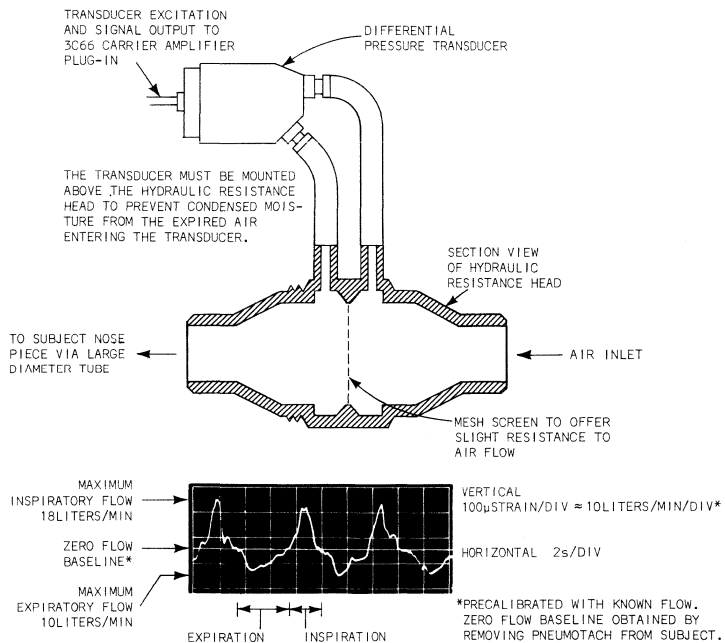


Fig. 9-6. Pneumotach air flow transducer.

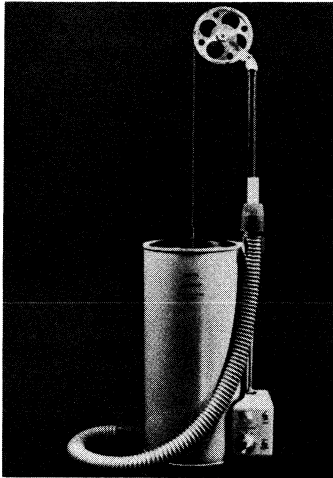
Commercial pneumotachs usually provide a calibration of the differential pressure produced per 10 liters per minute flow. This information, when related to the sensitivity of the pressure transducer, can be used to provide a calibration factor, as indicated in Chapter 18. A typical flow pneumogram is shown in Fig. 9-6. A zero base line was added to this display to provide zero flow reference by simply removing the subject from the pneumotach system and adding an additional sweep to the previously stored pneumogram.

Respiratory air flow measurement is frequently used to estimate a subject's respiratory function. Flow measurement also allows respiratory volume to be easily obtained.

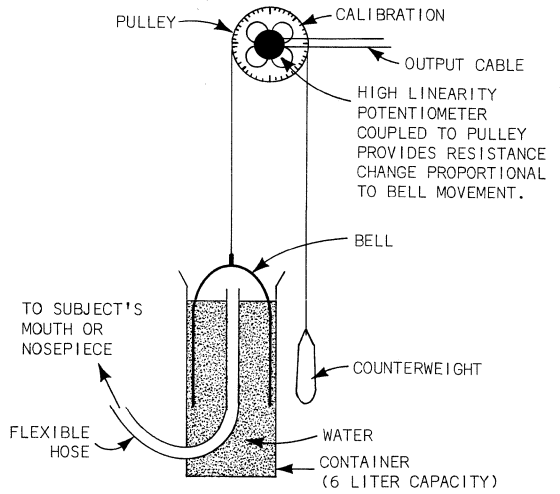
9.4 RESPIRATORY VOLUME

Since air flow is simply a measurement of volume per unit time, respiratory air flow information may be integrated to provide respiratory volume. Such a system is shown in Fig. 9-7. A pneumotach and differential pressure transducer produces an output proportional to respiratory flow as discussed in the previous section. To simplify the instrumentation requirements, the pressure transducer is operated from a ≈ 13 milliamperes DC source (ocilloscope Calibrator set to 40 V DC) and the resulting output amplified by one of the operational amplifiers in the Tektronix Type 3A8 Operational Amplifier. The amplified flow signal is then integrated using the second operational amplifier in the 3A8 unit. The output from this operational amplifier is thus an indication of respiratory volume. With the above system, the resistors and capacitors associated with the operational amplifiers can be selected from the front panel of the 3A8, with the exception of one of the 10,000 ohm resistors in the amplifier which must be added between the appropriate terminals on the front of the 3A8.

Minute respiratory volume may be measured by modifying the above procedure. Minute respiratory volume is the amount of air that a subject inhales in a one minute period. It may be measured by integrating *inspiratory flow* only, over a one-minute period. Referring to Fig. 9-7, the output of the second operational amplifier in the 3A8 unit will register minute volume if the flow signal is coupled to the integrating circuit via a diode and the integrator is gated ON for a one-minute period.



A COMMERCIAL SPIROMETER



SPIROMETER - FUNCTIONAL DETAILS

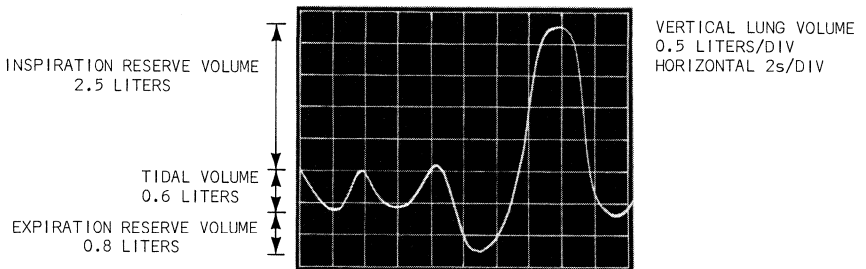
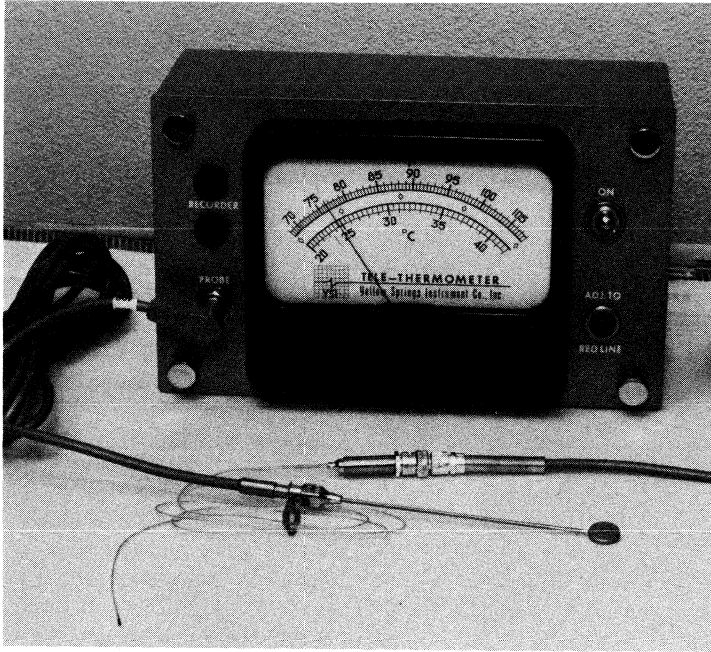


Fig. 9-8. A spirogram obtained with a spirometer.

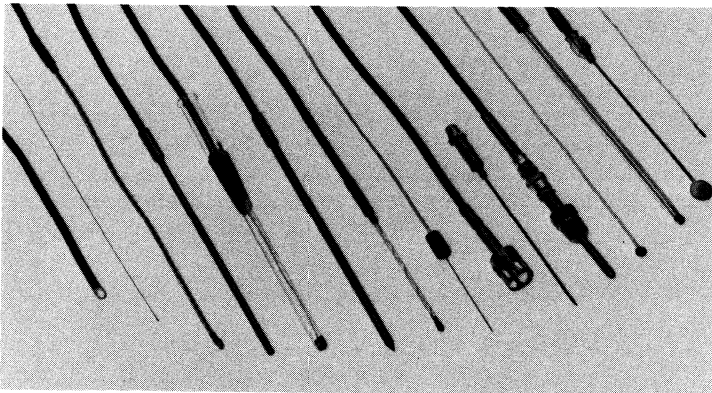
conventional
spirometer

A more conventional spirometer is shown in Fig. 9-8. Referring to this device, inspiration and expiration raises and lowers a counterbalanced bell located in a container full of water. Movement of this bell is transferred to a pulley whose periphery contains a calibration of bell displacement which is, of course, related to bell air volume. Respiratory volume may be read directly from this calibrated pulley. This pulley may also be coupled to a high-linearity potentiometer and the resistance change in this potentiometer used to indicate respiratory volume. The ≈ 13 milliamperes DC output from a 564B Calibrator is used to provide a constant-current source for this potentiometer; the output voltage will then be proportional to the changing resistance. A spirogram recorded with this system is also shown in Fig. 9-8. A spirometer is inherently a heavily damped device, containing appreciable hysteresis, so small subtle changes in inspiration and expiration volumes are not recorded with this device.

Fig. 9-2 shows theoretical changes in total lung volume with inspiration and expiration. Neither the integrating pneumotach or the spirometer can show total lung capacity as neither of these instruments have the ability to measure the residual volume of the lung. This must be measured by gas dilution techniques, which are beyond the scope of this book.



YELLOW SPRINGS INSTRUMENT CO. TELE-THERMOMETER WITH AN OUTPUT FOR A RECORDER.



VARIOUS THERMISTOR PROBES AVAILABLE FROM YELLOW SPRINGS INSTRUMENT CO.

Fig. 9-9. A commercial thermometer for use with thermistors.

9.5 TEMPERATURE

thermometer
using a
thermistor

In most cases, temperature does not vary at any appreciable rate and, thus, may be displayed using a moving coil meter. Such a thermometer, using thermistors in conjunction with a moving coil meter, is shown in Fig. 9-9. For some applications, however, the meter display is inadequate and thus an oscilloscope or chart recorder is required. The meter shown in Fig. 9-9 provides an output for use with an oscilloscope. Perhaps the main application where temperature does vary rapidly is in the recording of respiration, as covered earlier in this chapter.

While other temperature sensing devices, such as thermocouples, are feasible they are rarely used and small thermistors are used almost exclusively for temperature detection. These thermistors are available in a wide variety of sizes and mounting styles as shown in Fig. 9-9.

ELECTROENCEPHALOGRAPHY

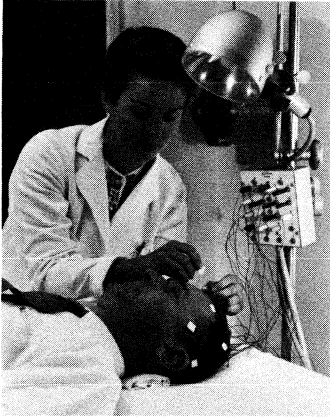
The following is necessarily a short and therefore somewhat incomplete survey of the EEG and EEG measuring techniques. Interested readers will find W. Grey Walter's text, *The Living Brain*, 1953, Duckworth and Co., invaluable for further study.

Electroencephalography, conveniently abbreviated EEGy, is the study of the electrical activity of the brain. Usually this activity is recorded from electrodes placed on the scalp, although some relatively rare diagnostic procedures require electrodes on or beneath the cerebral cortex. The EEG has been known for some 40 years and has made many contributions to man's knowledge of brain function. For reasons which are examined later, it has been of greater help to neurology (the study of brain function) than to psychiatry (the study of mental processes).

10.1 THE CHARACTERISTICS OF THE NORMAL EEG

The EEG of a normal adult human, normal being used in the everyday sense of the word, is relatively easily described. When the subject is relaxed, but not drowsy, a relatively smooth oscillation, whose frequency is seldom less than 8 Hz or more than 13 Hz, can be recorded from the area of scalp immediately over the occipital lobes. (Refer to Chapter 4, Fig. 4-2.) Typically this oscillation, the α rhythm, has an amplitude of 50 μ V peak-to-peak, although in rare subjects it may be twice this amplitude and in about 10% of the population it is absent or very small. This rhythm is responsive to mental activity; in most subjects attempting a task such as mental arithmetic will attenuate or abolish it.

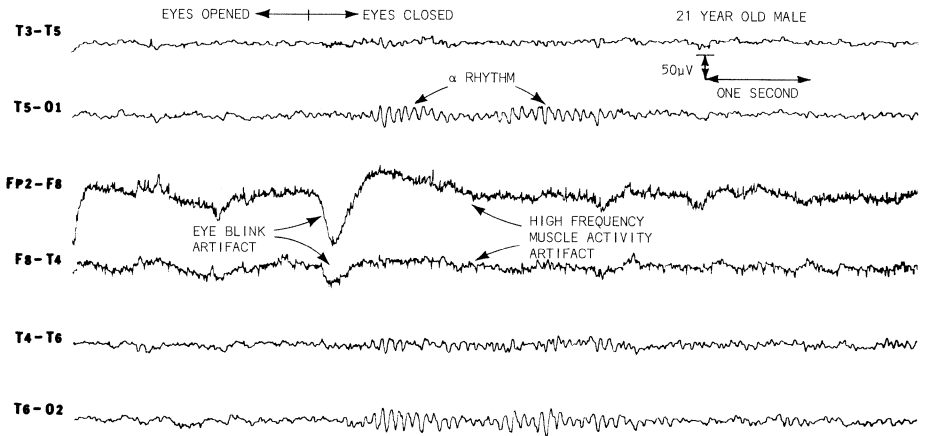
α rhythm



(A) ELECTRODES ARE APPLIED TO THE SCALP AND PLUGGED INTO THE JUNCTION BOX.



(B) A SWITCH SELECTOR ALLOWS THE DESIRED ELECTRODE CONFIGURATION TO BE CHOSEN.



(C) A SEGMENT OF THE RECORD OBTAINED SHOWING SIX OF THE SIXTEEN CHANNELS RECORDED.

Fig. 10-1. A typical adult EEG from a normal subject.

recorder Most EEGs are recorded using multi-channel ink writing oscillographs, as shown in Fig. 10-1, historically because they were widely available to physiologists. If some more sophisticated method of display is used it is found that more than one generator is involved, that there are generally several different frequencies, that there are differences in responsiveness between the cerebral hemispheres and that often the frequency of the signal measured in a transverse plane is consistently different from that observed with laterally placed electrodes.

frequency since the basic frequency of the EEG varies greatly with different behavioral states. To assist in the EEG analysis, the normal frequency range of the EEG (.5 Hz to 30 Hz) has been subdivided into five bands:

delta (δ).	0.5 Hz - 4 Hz
theta (θ).	4 Hz - 8 Hz
alpha (α).	8 Hz - 13 Hz
beta (β).	13 Hz - 22 Hz
gamma (γ).	22 Hz - 30 Hz

Various techniques for signal display are discussed later in this chapter. Although the α rhythm is the most prominent activity in the EEG of healthy adults, it is not seen in very young children and its absence does not indicate a lack of mental health or any deficiency in intelligence.

artifacts A segment of an EEG record from a normal adult male is shown in Fig. 10-1. Six of the 16 channels commonly recorded by an EEG instrument are shown. The tracing is read from left to right. Initially, the subject's eyes were open but after about 2.5 seconds he was asked to close them. The large downward deflection in leads FP2 - F8 and the smaller one in F8 - T4 are the "eye blink artifact". The α rhythm can be seen in the occipital channels T5 - O1 and T6 - O2 after the eyes were closed. Although the subject was completely normal, the α rhythm is somewhat smaller and less persistent than usual. The high frequency component in the two middle tracings is an artifact due to muscle activity and is not from the brain. The EEG shown in Fig. 10-1 represents only about eight seconds of recording, however, in practice, a recording may be maintained for an hour or more, producing a vast quantity of information for analysis.

10.2 EEG ELECTRODE CONSIDERATIONS

From an engineering standpoint the design of an EEG instrument and its accessories (electrodes etc.) is nowadays a routine matter requiring little more than ordinary care and attention to detail. As is so often the case in electronic design, the overall system limitations are almost all in the input devices, (the electrodes) which interface the equipment to the subject, and in the methods of storing the output data.

input
electrodes

The input electrodes are the most critical components of the recording chain. To be of use for routine EEG recording they must be small, be easily affixed to the scalp with minimal disturbance of coiffure, cause no discomfort and remain in place for extended periods of time. They must also have some fairly rigid electrical specifications if the signals are to be recorded with acceptably low levels of distortion.

signal
sources

We have noted that many EEG signals are of microvolt levels and it should be remembered that the signal is arising not at the scalp but in the cerebral cortex which is separated from the scalp by the cerebral spinal fluid (in which the brain is suspended) and by the skull. Parenthetically, we should note that engineers often suppose the skull to be an insulator because they usually see it dried and mounted. The living brain, however, is encased in living bone which is well permeated with conducting fluid. The amplifying system thus sees signals which arise in generators which have large, complex and variable source impedances. There may be large electrode offset potentials of the order of many millivolts developed between the electrode and the scalp unless a suitable electrode material is used. The high common-mode rejection ratio of the modern EEG amplifier will cancel the common-mode part of this signal but in practice small movements of the subject's head can cause substantial variations in the standing potential and if these are different in each lead they will of course appear as differential signals.

line interference

A further cause of problems in EEG recording is the presence in the modern clinic of many pieces of line operated equipment so that there are substantial magnetic and electric fields at the line frequency. The CMMR of the amplifier can, in principle, reduce these signals to insignificance but only if the entire system, including the electrode impedances, is balanced with respect to the common (ground) point on the amplifier. Thus electrode resistance must be reduced as far as possible; with good technique interelectrode resistances of 1 - 2 k Ω can be obtained. The alternative technique of reducing the line interference by the use of a shielded cage is not generally satisfactory since the degree of physical isolation it entails can be an emotionally upsetting experience, especially for a child. A relaxed subject is a necessity if good recordings are to be obtained.

electrode construction and connection

The most widely used electrodes are small silver pads electrolytically coated with silver chloride and attached to the scalp with a quick drying adhesive, usually collodion. A harness of rubber straps is also often used to hold the electrodes in place. Before the electrodes are applied, the scalp area is degreased and cleaned with alcohol and the surface resistance reduced by the use of a conducting paste. These electrodes are satisfactory for most recordings in the range 1 - 60 Hz. If, however, the low frequency limit is to be extended, which is the case in some research applications of the EEG, then electrodes which more closely approximate truly nonpolarizable electrodes, such as Tektronix Ag/AgCl electrodes must be used. Electrodes are generally placed at standard locations on the scalp to facilitate communication between electroencephalographers. These positions, with their usual designations, are shown in Fig. 10-2. The usual abbreviations are: F = frontal, T = temporal, C = central, P = parietal, O = occipital.

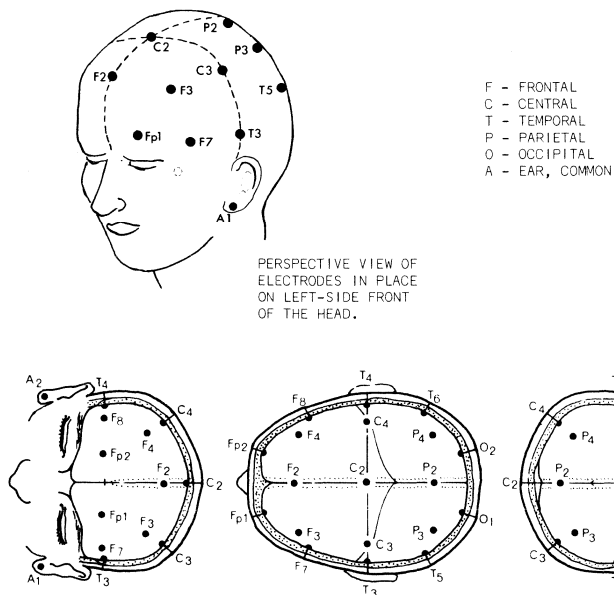


Fig. 10-2. EEG electrode positions.

10.3 EEG RECORDING INSTRUMENTS

Turning now to the output end of an EEG recording system, multi-channel ink-writing oscillographs are used for many reasons. The recording material is relatively cheap, the record is available for inspection as it is being written, the electroencephalographer can quickly flip through a long recording and obtain an "eyeball" impression of its contents and he may study interesting or complex parts of the record for as long as is necessary. Other media such as magnetic tape do not possess these properties and have not become popular in the routine clinical laboratory. If visual analysis is to be supplanted by computer or other automated data-processing techniques, then the written record must be supplemented by a magnetic recording or a curve reader (e.g., the multichannel high-speed curve reader described by Barlow in 1968) must be used. The frequency response of most EEG systems is limited by the characteristics of the recorder to something of the order of 60 Hz but this is adequate for most clinical purposes.

EEG recording systems are usually self-contained units consisting of electrode switching networks, high-gain differential amplifiers and graphic recorders. Multichannel recording is almost invariably used, the number of channels ranging between 6 and 32 with 8 or 16 channels being the numbers preferred for routine work.

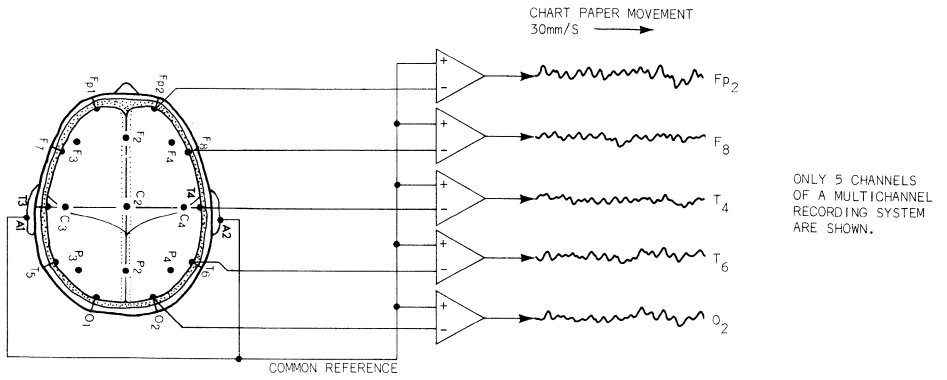
multichannel

A multiplicity of electrodes is affixed to the scalp as shown in Fig. 10-2 and the recording channels are connected to them via a switching network. The amplifiers are invariably designed to accept differential inputs and their design is usually optimized for low noise and good common-mode rejection. The low-frequency response usually extends to about 0.1 Hz, however the high-frequency response need not be in excess of about 100 Hz due to the limited high-frequency response of the graphic recorder following the amplifier. Since most EEG activity occurs below 50 Hz, a notch filter tuned to line frequency is often included in EEG instruments to minimize line frequency interference, but its use is strongly discouraged except as a last resort.

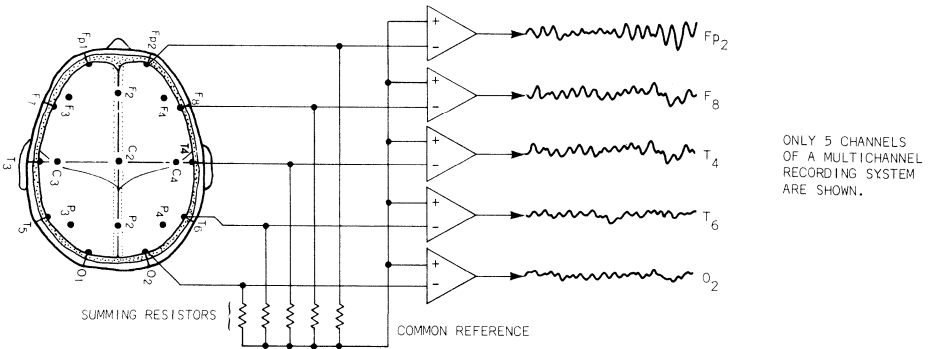
line
frequency
filter

sensitivity

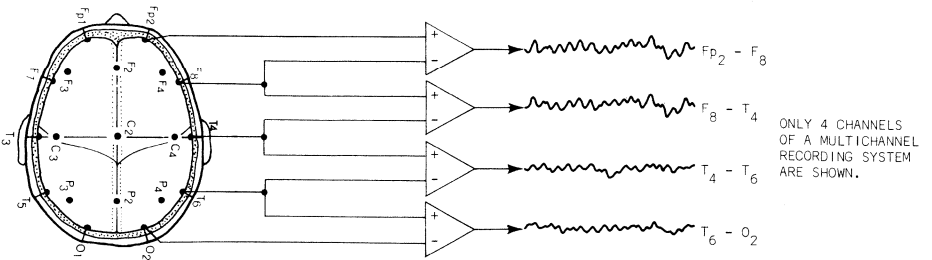
Although the gain of most modern EEG instruments is stable to within a few percent, a 50 μ V squarewave calibrator may be included in the instrument. Although the sensitivity of the amplifier may be adjusted to suit particular subjects, the electroencephalographer rarely changes the sensitivity during the recording of an EEG. He usually selects a gain that makes the initial record "look right" and uses the same gain throughout all phases of recording the EEG. Since the electroencephalographer is concerned with relative amplitudes between the channels, it is necessary that each channel has the same sensitivity. It is desirable to standardize on both sensitivity and paper speed to achieve aspect ratio consistency, which allows comparison with other EEG's recorded from other subjects. No firm standard exists here, however many workers prefer a sensitivity of about 7 mm per 50 μ V for adult subjects, a somewhat lower sensitivity for children and a somewhat higher sensitivity for the aged. The range of sensitivities used is usually within the range 4 mm per 50 μ V to 15 mm per 50 μ V.



(A) UNIPOLAR EEG RECORDING CONFIGURATION



(B) AVERAGE EEG RECORDING CONFIGURATION



(C) BIPOLAR EEG RECORDING CONFIGURATION

Fig. 10-3. EEG recording modes.

Most EEG instruments provide auxiliary outputs from all amplifiers to be used with other equipment, such as oscilloscopes, tape recorders or display devices. A paper speed of 30 mm per second (25 mm per second in some instances) is often used for electroencephalographic recording. Many EEG instruments may also add time markers to the EEG recording on a separate channel and may also incorporate electrode contact resistance measurement facilities.

10.4 EEG RECORDING MODES

Three modes of recording are used in the routine EEG laboratory as shown in Fig. 10-3. They are known as unipolar (often improperly called monopolar), averaging reference and bipolar recordings.

unipolar
mode

In the unipolar mode one electrode is common to all channels (Fig. 10-3A). Ideally, this common electrode is regarded as electrically inactive, however in practice, electrical activity near this electrode will appear in all channels and invariably there are problems in selecting the site for this common electrode. The ear, or both ears connected together, are sometimes used as being generally close to regions of the brain with little on-going electrical activity. If a subject has a localized discharge, for simplicity we will assume a spike discharge, then successful localization of the spike will be dependent on its amplitude in the various channels. With some loss of scientific vigor we may say that the amplitude will be greatest in the channel with its active electrode nearest the source of the spike. If the common electrode is near the spike focus, localization is either not possible or very ambiguous. Although one electrode is common to all channels, to reduce interference and artifacts it is desirable to not ground this common electrode and a separate ground electrode is often connected between the subject and the instrumentation ground.

average
electrode
mode

In the average electrode system one input lead of all amplifiers is taken to the common point of a summing network in which equal (high) resistors are taken to each electrode (Fig. 10-3B). The recording will now indicate deviations from the mean instantaneous potential of the electrode system and thus an isolated feature (e.g., the spike) will, if it is sharply localized, stand out in one, or at worst, a small number of channels. This recording mode can be loosely compared with the recording configuration used for unipolar ECG's as described in Chapter 5, Sections 5.4 and 5.7. A resistive summing network is used to create a common point in each case.

bipolar
mode

In the bipolar mode, the channels are connected in series between electrode pairs (Fig. 10-3C). By noting the change in the recorded EEG between these electrode pairs, very sharp localization of discharges is possible. The electrode immediately over the spike generator will cause a positive deflection in one recording channel and a negative deflection in the adjacent recording channel so that the electroencephalographer will see an apparent 180° phase difference between them. This "phase reversal focus" is accepted as the most reliable means of localization of discrete phenomena.

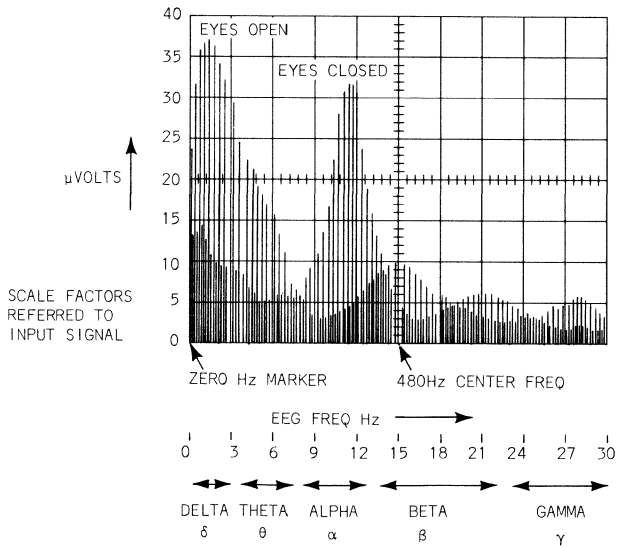
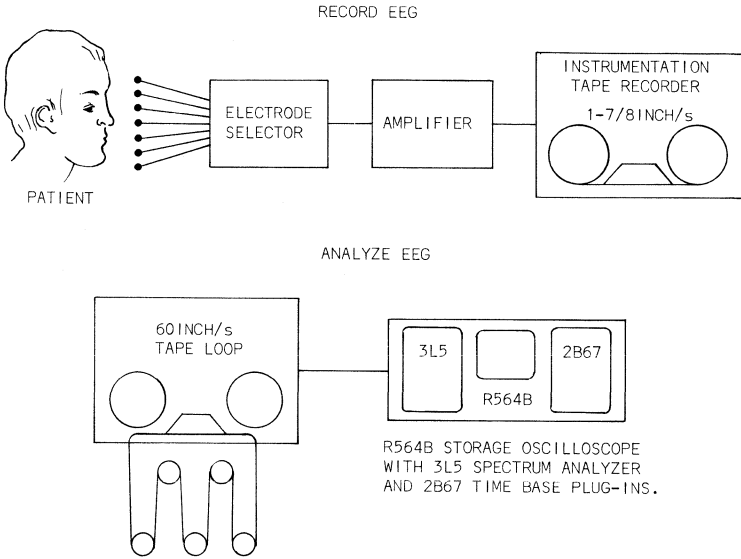
10.5 UNUSUAL EEG DISPLAY MODES

The voltage/time graph was used originally to display the EEG because oscillographs were available to the electrocardiographer. There is no certainty that these are the optimum ordinates to use in studying the EEG and a number of other techniques have been proposed. Some of these give a form of map-like presentation and allow the potential gradient over the head to be studied either in a "snapshot" fashion as used by Remond or as a synoptic display over a short time interval (Shipton). Such devices,

generally speaking, are good indicators of change in the EEG activity but poor for quantification, especially in terms of amplitude. It can be plausibly argued that amplitude is not the parameter of greatest interest and that what we are really concerned with is the way that the brain handles sensory data in terms of their temporal or spatial succession. The use of more sophisticated methods of display is increasing because the availability of on-line, real-time digital computers which have permitted data transformation in a number of interesting ways.

spectrum analysis

A spectrum analysis system has been occasionally used in research applications to present the EEG using amplitude/frequency coordinates. Most conventional sampling-type instrumentation spectrum analyzers are unsuited for direct EEG analysis as their low-frequency performance characteristics are inadequate and they do not lend themselves to the analysis of continuously changing data such as an EEG. The most desirable form of spectrum analyzers is a "real-time spectrum analyzer," however, such instruments are inherently rather expensive. Frequency analysis is rarely used as a clinical procedure; it masks much useful information which a human operator, using our superb pattern recognition abilities, can see at a glance. Mathematically, the difficulty of spectrum analysis of an EEG is that it is not time-invariant for a period long compared with the lowest frequencies present (refer e.g. Broadman & Tukey *Measurement of Power Spectra*).



EEG RECORDED AT 1-7/8 INCH/s, REPLAYED AT 60 INCH/s
"SPEEDUP" IS $32 \times$ CENTER FREQ = $15 \times 32\text{Hz} = 480\text{Hz}$

Fig. 10-4. EEG spectrum analysis.

tape
recorder

Spectrum analysis of the EEG has been performed with conventional instrumentation and sampling-type spectrum analyzers by utilizing an instrumentation tape recorder as indicated in Fig. 10-4. With this system a standard instrumentation tape recorder operating at 1 7/8" per second records the EEG and a small section of this recorded EEG is formed into a tape loop. This tape loop is then played back at 60" per second to effectively increase the frequency of the recorded information and to allow the use of a sampling-type spectrum analyzer, such as the Tektronix Type 3L5 Spectrum Analyzer in conjunction with a Tektronix Type 564B Storage Oscilloscope. The simulated CRT display shown in Fig. 10-5 represents two separate spectrum analyses of two separate tape loops, one recorded with a subject's eyes open and one recorded with a subject's eyes closed but with the subject awake. This display clearly shows the predominance of alpha activity with the eyes closed and the shift from alpha predominance with the eyes open.

10.6 INTRA-CRANIAL ELECTRODE PLACEMENT

In some diagnostic procedures the EEG recording electrodes are placed directly on the exposed surface of the brain. Under these conditions the output voltage will be considerably greater than the voltage obtained with normal EEG electrode placement, thus the gain of the recording instrument must be correspondingly reduced. Standard EEG electrodes cannot be used under these conditions as they are nonsterile and physically unsuited, thus special electrodes are used.

insulated
needle
electrodes

During neurosurgery insulated needle electrodes are often used to place an electrode deep within the subject's brain. These "deep electrodes" may consist of a needle insulated over its entire length with the exception of a small area at the tip or they may consist of concentric needles of varying length to effectively provide many electrodes at regular intervals along the length of the needle.

10.7 APPLICATIONS OF THE EEG

The EEG is primarily used in clinical neurology for partially assessing a subject's neurological state. Such applications are covered in more detail in the following section when discussing the abnormal EEG.

anesthetic level

As well as its utility in the neurological clinic, there are other uses for the human EEG. Brain cells are, for example, affected by anesthetic agents and the EEG is a sensitive indicator of the depth of anesthesia. Some workers indeed have used the EEG signal in a closed-loop controller to keep a constant anesthetic level.

monitoring during surgery

In many surgical procedures involving the heart, the ECG waveform cannot be monitored, thus the EEG signal is used as an indication of subject well being. With these procedures the verification of death can no longer be related to the activity of the cardiac system, thus the presence of EEG activity is, in part a useful indicator. EEG monitoring during surgery does not require the use of multichannel EEG instrumentation. The EEG signal can be monitored using a single channel recorder or, more commonly, by using an oscilloscope. Many surgical monitors, including the Tektronix Type 410 Physiological Monitor, designed primarily for cardiac monitoring, include EEG monitoring facilities. The 410 monitor simply requires two electrodes placed on the head over the occipital regions for EEG recording and one ground electrode placed anywhere on the subject.

alertness monitor

The EEG is also a very subtle estimator of the differences between sleep and wakefulness. Much of our present knowledge of sleep phenomena - and sleep is much more complex than it seems at first sight - we owe to the EEG observation of sleeping subjects. A number of states can be distinguished. For those who must be alert at specified times during a long task, piloting a spacecraft is a case in point, the EEG can be made the basis of a reliable "state of alertness" monitor.

stimuli
responses

Recently there has been great interest in the slow (circa 0.1 Hz) potential changes in the EEG. Modern amplifier techniques and nonpolarizable surface electrodes have established that these shifts in the DC potential are associated with voluntary responses to stimuli. It is very probable that such studies will extend the use of EEG techniques into the realm of psychophysiology; already a number of tentative relationships between mental state and variation of the slow "expectancy waves" have been established.

10.8 THE CHARACTERISTICS OF THE ABNORMAL EEG

epilepsies

In describing the EEG as it is seen in various disease states, it is important to remember that very few single clinical tests are by themselves sufficient to make a diagnosis. Thus, the EEG is only one of many procedures used by the clinician in assessing the neurological state of a subject. The most common condition in which the EEG is valuable is epilepsy. Strictly speaking, we should refer to "the epilepsies" since many varieties are found. The EEG is of great help in forecasting the outcome of an epileptic illness and is valuable in establishing the optimum course of treatment.

brain
injury

For example, injury to a specific region of the brain can leave a permanent scar on the cerebral cortex. Such scar tissue is electrically inert but has an irritative effect on nearby healthy cortex. The EEG will often show a localized spike discharge and will suggest, among other possibilities, surgical removal of the damaged tissue.

inborn
epilepsy

When epilepsy is inborn, and some forms of this disease are hereditary, the abnormal electrical activity generally contains signals at many frequencies in the range 1 - 50 Hz. There is no consistent phase relationship between the various components so that the EEG presents, to the eye, a "noisy" appearance. The signals are generally a good deal larger than the α rhythm and usually cannot be localized to any specific region of the brain.

"petit mal" In the so-called "petit mal" epilepsy, in which the manifestation of the illness is often a transient loss of consciousness or some automatic motor behavior, the signals are wideband but have remarkably consistent phase relationship between each component. The signal is thus seen as a regular pattern in which a sharp spike appears superimposed on a smooth low frequency (1 - 3 Hz) wave. Although many hypotheses have been advanced to account for the remarkable phase consistency seen in this "spike and wave" phenomenon, none are entirely convincing and all are outside the scope of this chapter.

tumors When the brain is invaded by some forms of tumor, a considerable portion of the active nervous tissue may be displaced by the electrically inert new growth. If this is very large, its presence can be inferred from the absence of organized electrical activity from the region of the tumor. Usually, however, a tumor large enough to be associated with a detectable area of "electrical silence" is large enough to manifest itself in other ways so that this technique of tumor detection is of limited practical value. There are, however, other ways in which the EEG can help in the diagnosis of brain tumors at a much earlier state of their development. The expanding new growth can interfere with the blood supply to neighboring areas and the consequent malfunctioning of nerve cells around it manifests by a large, slow discharge - the δ rhythm. There may also be significant differences in the electrical activity in the affected hemisphere, perhaps as a result of interference with the internal communicating pathways within the brain. The extent to which the EEG is of value in tumor localization depends on many factors. Some of the more important are the rate at which the tumor is growing, its special relationship to the recording electrodes and the skill of the electroencephalographer.

The conditions noted above are those in which the EEG is most frequently used. In other circumstances such as certain toxic conditions and some psychological states, the EEG can add to the overall amount of clinical information and be of significant benefit to the subject and physician.

10.9 INTENTIONAL MODIFICATION OF A SUBJECT'S EEG

external
stimuli

Up to this point we have assumed that the electroencephalographer plays an entirely passive role and is content to study the brain in its normal physiologic milieu. In practice a number of techniques are used to increase the yield of meaningful information; some of these apply external stimuli to the subject and note their effect on the EEG. Because the excitability of various parts of the nervous system is critically dependent on the acid-base balance (and thus on the oxygenation of the blood) it is general practice to modify this balance by asking the subject to hyperventilate, that is to say breathe rapidly and deeply while at rest. In the normal subject this procedure produces a nominal slowing of the α rhythm and some small increase in the overall signal level. If, however, the subject is an epileptic, the record is dramatically changed to the extent that a seizure may be provoked. Some epileptics are markedly affected by lowering their blood sugar and for this reason many clinical records are obtained from fasting subjects.

Another important means of modifying the EEG is the use of rhythmic sensory stimulation. One or more of the senses is stimulated by brief repetitive stimuli; light flashes are the most commonly used partly because they are easy to generate and partly because the visual cortex is large and the source of the α rhythm. Sensory stimulation of this kind can emphasize latent abnormalities in the resting EEG and help in the interpretation of the tracing.

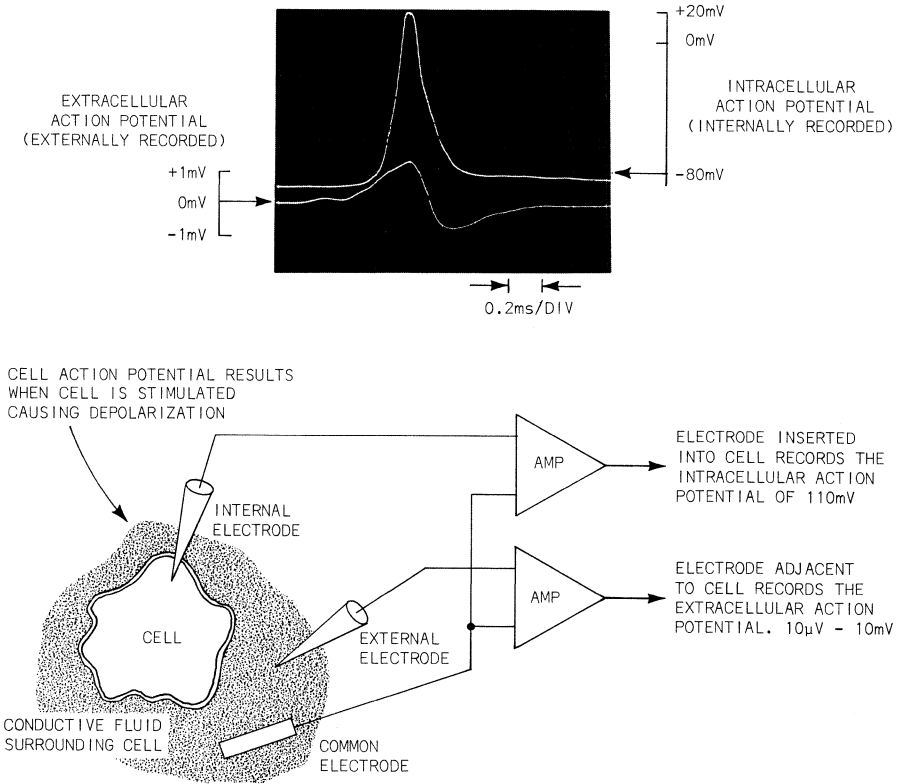


Fig. 11-1. Cell action potential — internally and externally recorded.

11

EVOKED CORTICAL RESPONSES

The electroencephalogram referred to in Chapter 10 is a measure of the over-all electrical activity of the brain with the subject essentially at rest. The electroencephalogram is probably associated with the computation process continuously active within the brain. Evoked potentials are also potentials generated within the brain, however these potentials result from a stimulus being applied to the body's sensory system and are localized to a particular area of the brain. These potentials are said to be "evoked" by the stimulus.

11.1 EVOKED ACTION POTENTIAL

As stated above, stimulation of a subject's sensory system produces electrical activity in a localized area of the brain. When attempting to analyze the effect of the stimulus, it is necessary to record the electrical potentials generated within *individual* brain cells by using intracellular recording techniques. These techniques are only used in research applications on nonhuman subjects. External electrodes will record the electrical activity of a cell, however, as many adjacent cells may also be producing electrical activity, the results obtained could not be attributed to any *particular* cell. Under certain conditions, particularly when recording from the spinal cord rather than from the brain, it may be possible to isolate individual cell activity using external electrodes to record the extracellular action potential. Fig. 11-1 shows the action potential generated by a *single* cell when recorded with both internal and external electrodes and shows the time and voltage relationship between these two recording techniques.

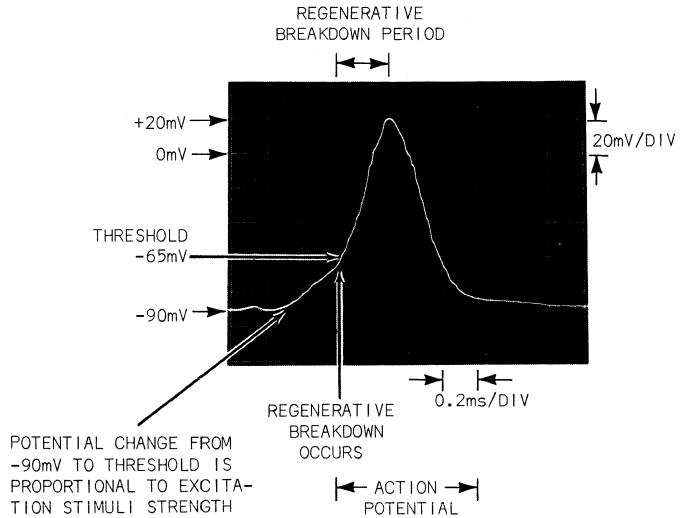


Fig. 11-2. Action potential showing threshold.

intracellular
action
potential

When recording the intracellular potential with an internal electrode, about 110 millivolts of signal is generated during a cell depolarization/repolarization process and this signal is known as the intracellular action potential. It is, however, possible to record lower amplitude signals generated within the cell as a result of excitation and inhibition stimuli as discussed in Chapter 4. The intracellular action potential shown in Fig. 11-2 is clearly preceded by a period where the cell is receiving excitatory stimuli which decreases the cell's resting potential at a linear rate until the cell threshold is reached, at which time the rate of change of potential increases, indicating regenerative depolarization. The cell subsequently repolarizes.

11.2 MICROELECTRODE TECHNIQUE

The preceding discussion covers intracellular recording and shows typical intracellular action potentials. In practice, intracellular recording requires highly specialized measurement techniques, and results similar to the action potentials shown in Fig. 11-1 and 11-2 are difficult to achieve.

When recording the intracellular action potential it is, of course, necessary to insert an electrode into the cell concerned. If the results obtained are to serve any practical purpose, it is also necessary that this electrode have a negligible effect on the characteristics of the cell concerned. It is thus desirable to use an electrode with dimensions much smaller than the dimensions of the cell concerned; requiring electrodes with tip diameters well under one micron (10^{-4} centimeters). These small electrodes are known as microelectrodes. Both glass microelectrodes, often referred to as micropipettes, and metal microelectrodes are available.

micro-
electrodes

The basic difference between these is that in the metal microelectrode the metal is in direct contact with the biological tissue whereas in the glass microelectrode an electrolyte is interspersed between the tissue and the metal electrode. Thus, metal microelectrodes have a lower resistance, but they polarize with smaller amplifier input currents, their resistance may increase, and they may develop unstable electrode offset potentials. Unless extreme precautions are taken, they are, therefore, unreliable for steady-state potential measurements.

metal

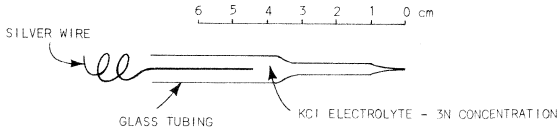
The glass microelectrode interposes an electrolyte between the tissue and the metal electrode which results in improved stability as the metal and the electrolyte can be chosen so that small, steady currents can pass through their junction without modifying the electrical properties of the electrode. The surface contact area between the electrolyte and the metal is large so that the current-carrying capacity of the electrode is substantial. The glass microelectrode is, therefore, usually preferred.

glass

Fig. 11-3 shows typical tip dimensions of a glass microelectrode, having an over-all tip diameter of about 0.6 microns with an internal diameter of 0.2 microns. The glass microelectrode is formed by heating special glass tubing and drawing it out over several stages of reduction. Although the tubing is reduced to less than a micron in diameter, it still remains hollow. Potassium chloride solution is introduced into the microelectrode as the electrolyte. As it is not possible to fill the microelectrode by pressure (surface tension) or by capillary action, boiling with or without reduced pressure is often used. Ideally, one would like to use an electrolyte within the microelectrode having the same concentration as the potassium chloride within a typical cell (0.1 Normal). It is, however, impossible to use an electrolyte of this concentration as the resistance of the microelectrode would be well over 1,000 megohms. It is, thus, common practice to use 2 or 3 Normal potassium chloride in the microelectrode. This electrolyte has a resistivity of 3.3 ohm centimeter which gives the microelectrode similar to that shown in Fig. 11-3 a typical resistance of 10 megohms. Special purpose microelectrodes having smaller tip diameters and/or using a less concentrated electrolyte may have a typical resistance of 100 megohms. As will be shown later in this chapter, special recording techniques are necessary to accommodate this extremely high electrode series resistance.

Fig. 11-4 shows a typical microelectrode inserted into a cell and the equivalent electrical circuit formed between this microelectrode and a common electrode elsewhere on the subject. This equivalent circuit can be further simplified to the microelectrode resistance with distributed capacity, and an RC load on the microelectrode formed by interconnection capacity, amplifier input capacity and amplifier input resistance. Since the dimensions of the microelectrode tip are extremely small, and as most of the microelectrode resistance is located within one millimeter of the microelectrode tip, this distributed capacities associated with the electrode resistance are less than one picofarad.

MICROELECTRODE OVERALL DIMENSIONS:



MICROELECTRODE TIP DIMENSIONS:

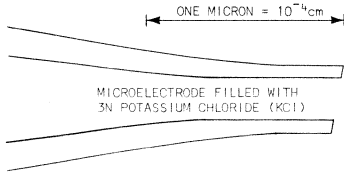


Fig. 11-3. Glass microelectrode geometry.

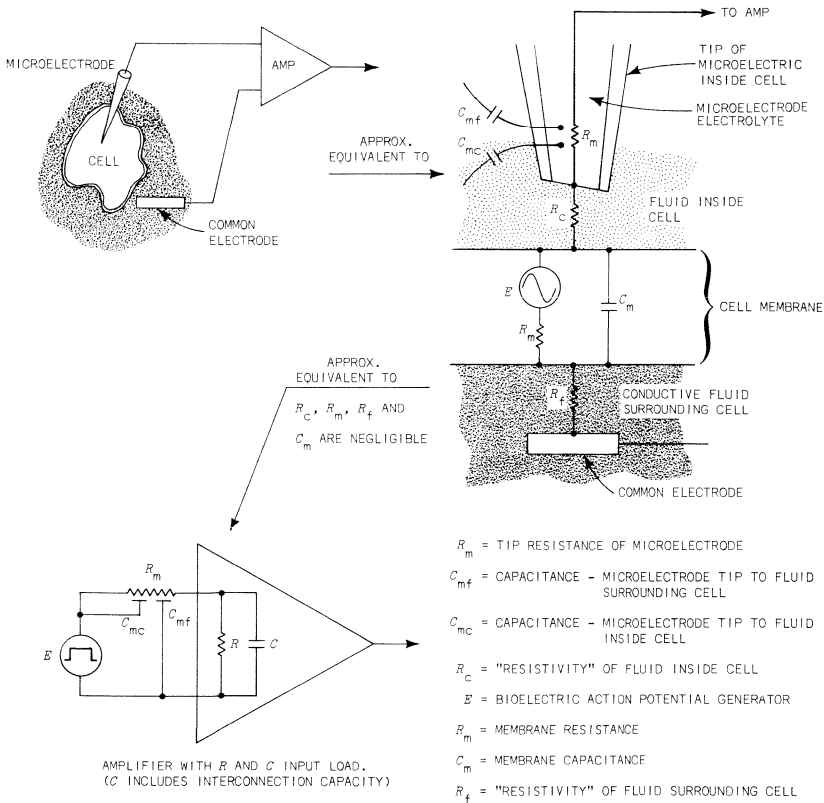
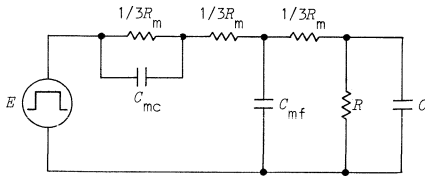


Fig. 11-4. Cell-microelectrode equivalent circuit.



EQUIVALENT CIRCUIT FROM FIG. 11-4.

TYPICAL VALUES:

$C_{mc} = 0.1\text{pF}$

$C_{mf} = 0.2\text{pF}$

$R = 1000\text{M}\Omega$

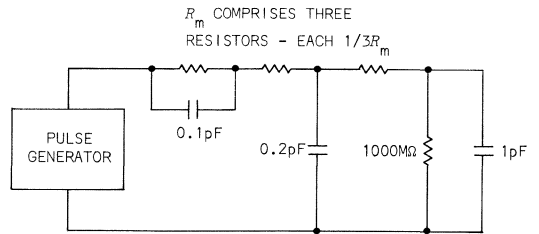
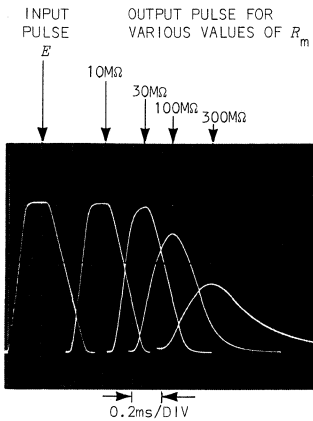
$C = 1\text{pF}$

$R_m = 10\text{M}\Omega$

$E = -90\text{mV TO } +20\text{mV}$

$t_r = 0.1\text{ms} \quad t_f = 0.2\text{ms}$

DURATION - 0.5ms



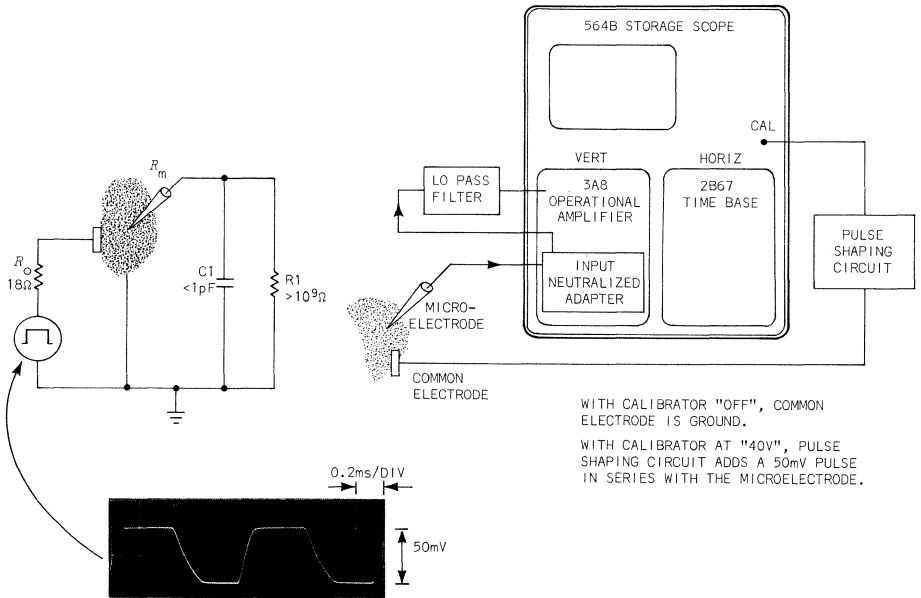
MICROELECTRODE EQUIVALENT CIRCUIT
MODELED WITH IMPEDANCES SCALED BY 10^3 .

Fig. 11-5. Microelectrode model and resulting pulse response.

circuit
determines
fidelity

Referring to Fig. 11-5, a typical microelectrode equivalent circuit is shown together with typical values for the impedances concerned. This equivalent circuit represents a low-pass filter and attenuator and it is necessary to know the characteristics of this circuit to determine the fidelity expected from the system when recording the action potential. This equivalent circuit can be modeled using a pulse generator in place of the cell and using discrete impedances in place of the microelectrode resistance and capacity. Fig. 11-5 shows the degradation that can be expected from an input "action potential" having a risetime of 0.1 millisecond and a falltime of 0.2 millisecond for various values of electrode internal resistance when using an amplifier having an equivalent input capacity of one picofarad and an equivalent input resistance of 1,000 megohms.

As can be seen from Fig. 11-5, a 100 megohm microelectrode used in the above system reduces the amplitude of the action potential by 20 percent and degrades its risetime and falltime. It is thus desirable to either reduce the resistance of the microelectrode to 30 megohms or perhaps even 10 megohms to increase fidelity or, alternatively, to increase the input impedance of the amplifier and interconnection network. Since decreasing the microelectrode resistance usually means increasing the size of the microelectrode, the latter alternative is preferable.



OUTPUT WAVEFORM FROM PULSE SHAPING CIRCUIT USED FOR SYSTEM CALIBRATION.

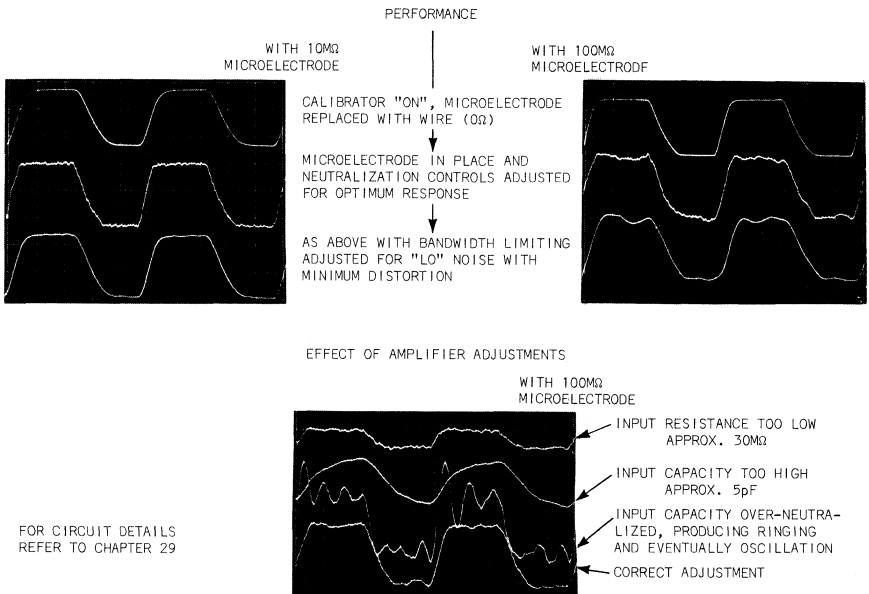


Fig. 11-6. An input neutralized amplifier and scope system.

11.3 INPUT NEUTRALIZED AMPLIFICATION

feedback
increases
input Z

The impedance load on the microelectrode can be increased by using active feedback amplification to provide input impedance neutralization. Careful adjustment of such amplifiers can achieve an input resistance in excess of 10,000 megohms with equivalent input capacities of the order of 0.1 picofarad. Fig. 11-6 shows a typical microelectrode system utilizing a Tektronix Type 3A8 Operational Amplifier plug-in unit operating in an "input neutralized" mode. This requires the use of an input neutralizing adapter in conjunction with the 3A8. Details of this adapter are given in Chapter 29.

system
calibration

This microelectrode recording system also incorporates a series calibration waveform. By switching the 564B calibrator from OFF to 40 V, a 50 millivolt, 1000 Hz, signal is added in series with the microelectrode. This signal has a controlled risetime of 0.1 ms and a controlled falltime of 0.2 millisecond and is used in conjunction with the two input neutralization controls (-R and -C) to check the amplitude and time calibration of the system. When the calibrator is returned to its OFF position, this signal source is essentially replaced by a short circuit. This calibration waveform and the response of the system for various settings of the input neutralization controls are shown in Fig. 11-6. In practice, the input neutralization should be adjusted to the point at which overshoot and ringing are reduced to less than 10 percent. Further reduction of the overshoot and ringing will degrade the risetime of the system.

amplifier
input
current

Although the dynamic input resistance of a neutralized amplifier may be above 10^{10} ohms, the system may have excessive amplifier input current and will thus not operate. Since the dynamic input resistance of any device is determined by the change in input current for an incremental change in input voltage; $R = \frac{\Delta V}{\Delta I}$; and as amplifier input current does not change with an incremental change in input voltage, this amplifier input current is not reflected by the input resistance specification of the amplifier and must be specified separately. Any neutralized amplifier should have an amplifier input current below 10^{-12} amperes.

The system shown in Fig. 11-6 has an amplifier input current of approximately 10^{-13} amperes. These extremely small currents are necessary when using the amplifier for intracellular recording as the current will cause potassium ions to migrate from the microelectrode into the cell. This upsets the intracellular concentration, which may change the cell resting potential. Small changes in resting potential due to extremely small currents cannot be avoided, and may not be significant. However, it is obviously desirable to maintain the current below the level required to cause cell depolarization.

tangential
noise

The noise in any neutralized amplifier is proportional to the degree of neutralization required; a tangential noise of several millivolts is not uncommon. "Tangential" noise is discussed in Chapter 19, Section 19.7. The system shown in Fig. 11-6, when correctly neutralized, produces a tangential noise of 1 millivolt when used with an electrode having a resistance of 10 megohms and 2.5 millivolts when used with an electrode having a 100 megohms resistance. This noise is insignificant when recording intracellular potentials as these potentials are in the order of 100 millivolts. For extracellular measurements however, the extracellular action potential may be as low as 10 microvolts and the amplifier noise will completely swamp any extracellular signals. This situation also arises when using intracellular recording at high sensitivity in an attempt to observe excitation and inhibition potentials. The signal-to-noise ratio can be substantially improved by a lowpass filter following the neutralized amplifier. Such a filter is included in the system shown in Fig. 11-6 and further details of this filter are given in Chapter 29. Even using a filter for extracellular measurements, the noise level obtained may still completely swamp the signal and, at best, will be extremely objectionable. Thus, particularly when working at sensitivities below 100 microvolts per division, some more effective noise reduction technique is almost always necessary.

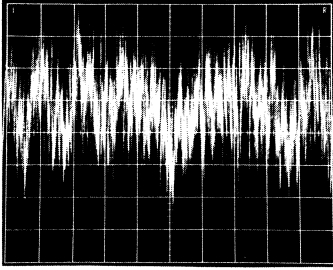
lowpass
filter

11.4 NOISE REDUCTION - AVERAGING

signal
averaging

A single response evoked by one stimulus is usually too small to be seen above the on-going EEG activity and the noise inherent in the system. However, if we assume that the response always follows the stimulus after a fixed delay, "signal averaging" computers can improve the signal-to-noise ratio. Many such signal averaging computers have been marketed in the past few years and, while they differ in detail, they all use the same basic principle.

Signal averaging extracts a wanted signal from a background of unwanted noise. It can only be used effectively if the desired signal, with its accompanying noise, can be generated a number of times, either periodically or aperiodically. In addition, a trigger pulse is required that has a fixed time relationship to the desired signal. The stimulus, or trigger pulse, initiates a scanning device which samples the signal at fixed intervals. These time-sequential samples are stored in discrete storage channels or memory locations, each channel collecting data over a small time segment. When the stimulus is repeated, the responses are added to the values stored at each location. During readout, each of the channels displays the sum of all the previous samples fed into it. If the background activity is random with respect to the stimulus, the time-locked response will add linearly with the number of samples (n) while the background will add only as the square root of n . If the stimulus is repeated 64 times (a fairly common number), then the response to noise ratio will have been improved by a factor of $\frac{n}{\sqrt{n}} = \sqrt{n} = 8$. Filtering to the maximum permissible extent should always precede the averaging device.



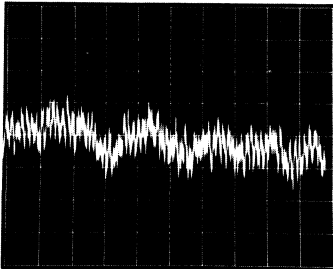
RAW SIGNAL OUTPUT FROM
NEUTRALIZED AMPLIFIER
VERTICAL 100 μ V/DIV (REFERRED TO INPUT)
HORIZONTAL 0.2ms/DIV



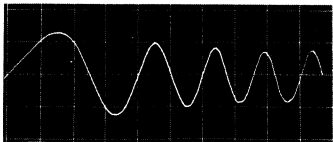
SIGNAL AVERAGER AND READOUT OSCILLOSCOPE

CONVENTIONAL SIGNAL PROCESSING

AMPLIFICATION WITH BANDWIDTH
LIMIT (10kHz) AND GAIN REDUCTION
(200 μ V/DIV REFERRED TO INPUT)



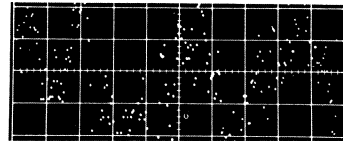
ORIGINAL SIGNAL WITHOUT NOISE.
IDEAL SIGNAL PROCESSING SHOULD
PRODUCE THIS WAVEFORM.



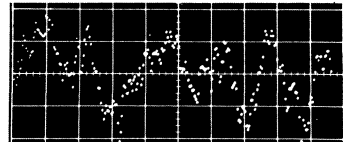
100 μ V/DIV 0.2ms/DIV

SIGNAL AVERAGING

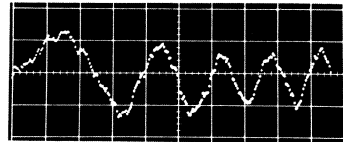
NUMBER OF
SWEEPS
OF AVERAGER



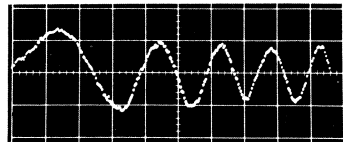
1 SWEEP



4 SWEEPS



64 SWEEPS



256 SWEEPS

100 μ V/DIV 0.2ms/DIV

Fig. 11-7. Signal to noise ratio improvement with averaging.

signal-to-noise ratio improvement

Typical signal-to-noise ratio improvement via averaging is shown in Fig. 11-7. The noise has a peak value about equal to that of the signal, equivalent to an RMS S/N ratio of about 4. Averaging is an ideal technique under these conditions. For greater noise levels the signal content is reduced, because the peak noise plus signal must still not overdrive the averager; difficulty then arises because the signal component alone is not being digitized to a sufficient number of bits to give a usable output.

triggering on signal

In many circumstances, averaging can be carried out in the absence of the trigger pulse if it is possible to use part of the response itself as the stable point. This is an aspect of averaging which is currently of great interest to physiologists.

averager characteristics

The resolution of the signal averager is limited by the number of addressable channels; commercial units may contain from 50 to 1,000 channels. The "speed" of the system is limited by the length of time required to store information into any one channel. Commercial averagers are able to sweep through all channels in less than 5 milliseconds, providing an effective sweep speed of 0.5 milliseconds per centimeter for the readout oscilloscope. Since the sweep speed is limited by the individual channel access time, the use of fewer channels can increase the effective sweep speed at the expense of resolution. The channel access time for most modern averagers is about 10 microseconds; however, some earlier averagers designed specifically for biological use have an access time in excess of 100 microseconds.

Most signal averagers contain the necessary circuits for averaging but require a separate X-Y oscilloscope to display the output information. Normally, calibration facilities are incorporated into averagers to allow the monitor to be calibrated for both voltage and time reference. This calibration is particularly important as the "level" of each channel in the averager varies with the number of sweeps used and all absolute amplitude information is, therefore, meaningless unless a calibration system is provided. Some time averagers are now available in which the *average* rather than the *sum* of the sweeps is displayed.

The signal averager relies on a trigger pulse being available having a known time relationship to the desired signal. When recording evoked potentials, it will be recalled that these potentials are generated in response to a stimulus being applied to the patient, this stimulus can be used as the trigger for the averager.

using
stimulus as
trigger

The averaging technique described above can introduce considerable error if time inconsistencies (jitter) exist between the trigger pulse and the evoked response. The response shown in Fig. 11-8 was produced after averaging over 256 sweeps, each sweep being initiated by light flashes at an approximate 2 second rate; thus 512 seconds or about 8.5 minutes are required to produce the response. During this long time interval, both the shape of the evoked response and the timing relationship between this evoked response and the light stimulus may vary and the simple averager will produce a response of a longer duration and shorter amplitude than will be the case if these time inconsistencies did not exist.

time
jitter

More sophisticated signal-averaging techniques, using digital computers, although inherently more expensive than the techniques described above, are somewhat more versatile as they also incorporate facilities to analyze statistically the results obtained. The Digital Equipment Corporation Lab 8 signal averager is typical of a computer-type signal averager.

digital
computer
averagers

Many other computational aids are in use to improve the quality of signal recovery. For example, the presence or absence of a rhythmic component in a signal can be determined by cross-correlating the signal from one amplifier channel with that in another. This technique is a sensitive indicator that a signal is present but does not give information about its amplitude nor necessarily about its phase or time relationship.

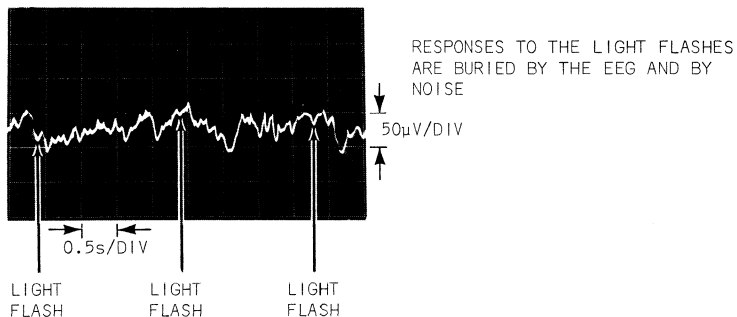
correlating
two inputs
to detect
signal

Although the signal averager is primarily intended to detect a wanted signal in the presence of unwanted noise, it can also be used to detect a wanted signal in the presence of other unwanted signals. When recording the EEG, for example, it is possible to detect small changes in the EEG when flashing a light at the subject.

averaging
evoked
responses
from
unwanted
signals

Using conventional amplification techniques, it is possible to detect the response of the brain to this light flash as this response is camouflaged by EEG activity produced within other parts of the brain. The signal averager can be used to detect this response by regarding the EEG as an unwanted signal which bears no time relationship to the stimulus. Fig. 11-8 shows a typical EEG recorded with a light being flashed at the subject at approximately two-second intervals. The EEG shown appears normal and it is almost impossible to detect the effect of the light flash. The same EEG signal, when analyzed by an averaging computer triggered from the light flash, clearly shows the electrical response of the brain to the light flash as seen in Fig. 11-8.

EEG RECORDED WITH SCALP ELECTRODES DURING REPEATED LIGHT FLASHES OF 1ms DURATION AT APPROX. TWO SECOND INTERVALS.



THE ABOVE WAVEFORM AFTER ANALYSIS BY A SIGNAL AVERAGING INSTRUMENT TRIGGERED FROM THE LIGHT FLASH.

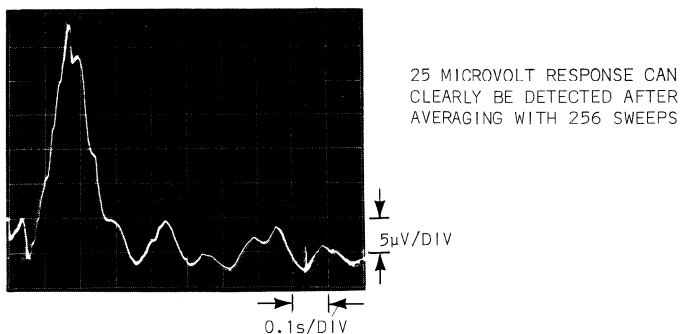


Fig. 11-8. Use of averaging to detect a response in the EEG.

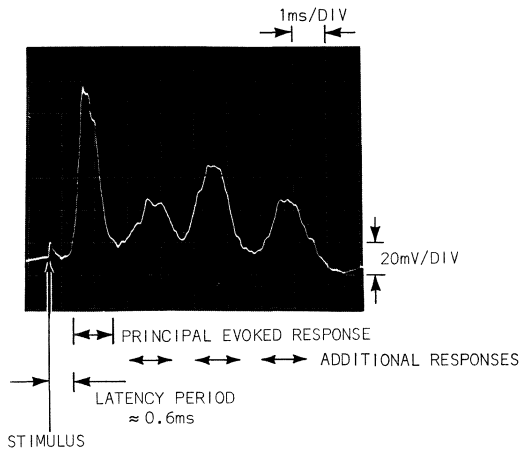


Fig. 11-9. Evoked response showing latency and additional responses.

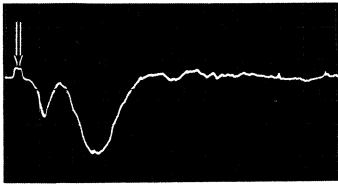
11.5 TYPICAL EVOKED RESPONSES

stimulus/
response
delay

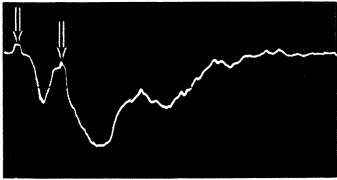
A typical evoked response obtained with a microelectrode on a laboratory animal is shown in Fig. 11-9. This response waveform is preceded by a small amount of stimulus artifact. Although it is desirable to reduce the level of stimulus artifact appearing in recordings, it is often undesirable to completely suppress it as it does provide some time relationship in the recorded evoked potential. The evoked potential begins sometime after the stimulus is received; this delay is known as latency and is approximately 0.6 milliseconds in the waveform shown in Fig. 11-9.

Typical evoked potentials obtained with an extracellular microelectrode with stimulation of the visual cortex of a laboratory animal are shown in Fig. 11-10. For the reasons covered above, stimulus artifact has deliberately been added to these recordings to provide a time reference. Fig. 11-10 shows the evoked response obtained for a single light flash and for multiple light flashes separated by various periods from one another. The 50 millisecond pair of flashes is seen by the subject as one, whereas two flashes are perceived for the 150 and 250 millisecond intervals. Unlike the evoked response shown in Fig. 11-9, these responses were obtained with extracellular electrodes and show the effect of many cells within the visual cortex; thus the "time scale" of the response is greatly increased as many cells are involved.

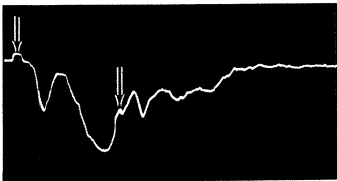
RESPONSE TO SINGLE AND DOUBLE FLASHES OF LIGHT. EACH FLASH IS 10 μ s DURATION. (ARROWS INDICATE LIGHT FLASH.)



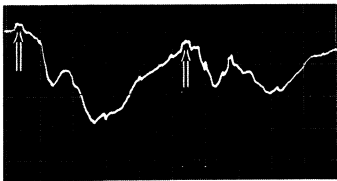
SINGLE FLASH
SINGLE RESPONSE OBTAINED



DOUBLE FLASH
50ms APART
SINGLE RESPONSE OBTAINED



DOUBLE FLASH
150ms APART
DOUBLE RESPONSE OBTAINED



DOUBLE FLASH
250ms APART
DOUBLE RESPONSE OBTAINED

→ | ← | 100 μ V/DIV
50ms/DIV

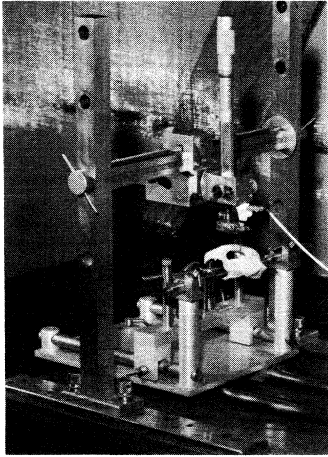
Fig. 11-10. Evoked potentials from the visual cortex.

11.6 STIMULATION

This chapter has previously discussed potentials produced as a result of stimulation, however the techniques used to provide this stimulation have not been covered. Stimulation is covered in detail in Chapter 12 when discussing the Electromyogram and Nerve Conduction. The principles applied to stimulating the peripheral nerves and muscles also apply to stimulating the cortex. It should be noted, however, that cortex stimulation may be carried out with the brain exposed and with stimulation electrodes spaced by only a few millimeters so as to provide extremely localized stimulation. Under these conditions, a greater percentage of the stimulation current passes through the cells of interest and, thus, the overall stimulus current required to produce a response is far lower than is the case when stimulating through skin and bone with electrodes placed many centimeters apart. Generally, when stimulating with the cranium opened, stimulators capable of outputs up to ten volts are adequate. Stimulus electrodes are normally needles or concentric needles -- the concentric needle provides stimulation between two "electrodes" less than one millimeter apart and also provides some degree of shielding to minimize the level of stimulus artifact appearing in the recorded response. Since electrical stimulation may create stimulus artifact potentials which camouflage the response, and since it appears to be desirable to stimulate via a sensory modality rather than to use direct stimulation, light stimulation or other sensory receptor stimulation is often used.

11.7 STEREOTAXIC INSTRUMENTS

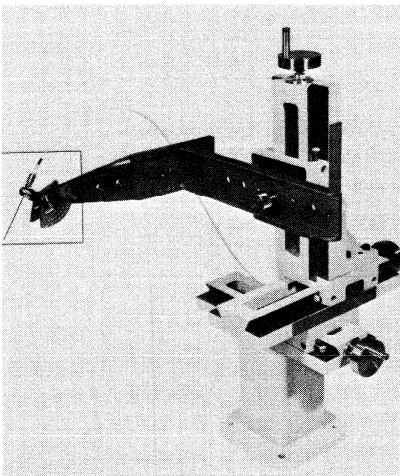
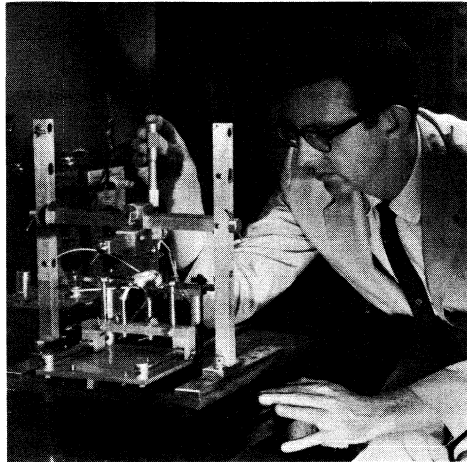
The instrumentation system shown in Fig. 11-6 simply depicts a microelectrode inserted into a specimen. In practice, when recording evoked potentials, the skull is held firmly in a jig and the microelectrode is positioned with a micromanipulator or stereotaxic instrument, usually with the aid of a microscope or X-ray techniques, as shown in Fig. 11-11. Although stereotaxic instruments are available commercially, many are constructed from precision microscope stages.



} STEREOTAXIC INSTRUMENT

} RIGID SUPPORT FOR ANIMAL'S SKULL

A TYPICAL STEREOTAXIC INSTRUMENT AND SKULL SUPPORT

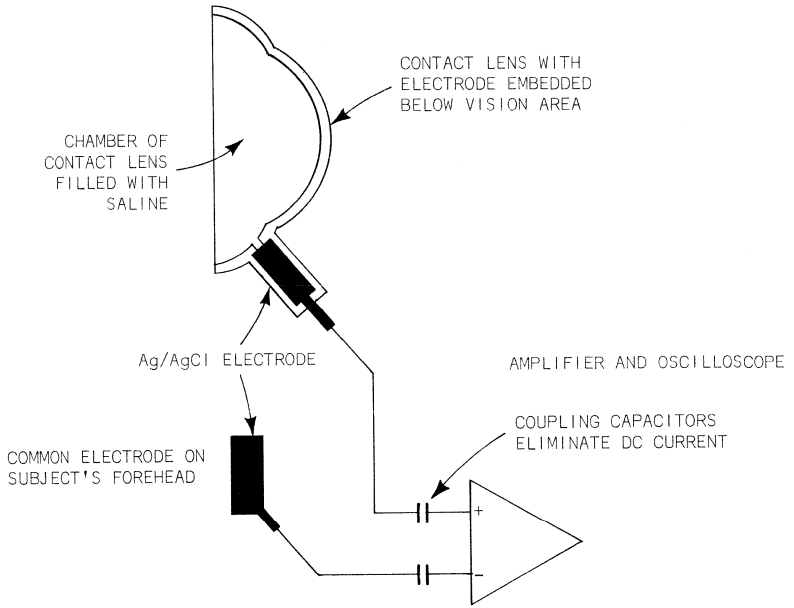


THE FRY* PROBE AND ELECTRODE POSITIONING UNIT

VERNIER SCALES ON THE X, Y AND Z AXES PERMITS POSITIONING TO 0.05mm

*INTERSCIENCE RESEARCH INST. CHAMPAIGN, ILLINOIS

Fig. 11-11. Sterotaxic instruments.



STROBE UNIT

Fig. 11-12. Contact lens ERG electrode.

11.8 THE ELECTRORETINOGRAM -- ERG

As stated in Chapter 3, it is almost impossible to detect the electrical activity produced by most of the sensory receptors on the body as they are too small and are dispersed. The notable exceptions to this are the potentials resulting from stimulation of the middle ear by sound waves and of the retina by light. If a bright light is projected into the eye, the retina will be stimulated and will generate action potentials known as the electroretinogram, or ERG, which can be detected by an electrode placed on the outside of the retina.

electrodes The human electroretinogram is recorded with the aid of a silver/silver-chloride electrode imbedded in a contact lens, and a "common" silver/silver-chloride electrode placed on the subject's forehead, as shown in Fig. 11-12. Application of the contact lens electrode is painless and it is thus suited to clinical use. The subject is placed in front of a light source which may simply be an incandescent light coupled to a photographic shutter or, more commonly, a stroboscopic light as shown in Fig. 11-12. The General Radio Company Type 1504 P4 Stroboscope is ideal for this application.

stimulus Stimuli are presented in the form of flashes of light and the responses, which are obtained by means of the contact lens electrode, are displayed on a medical monitor, such as the Tektronix Type 410 Physiological Monitor. The 410 is operated in the EEG mode with the "EEG electrodes" formed by the contact lens and common electrode. Careful technique is required in affixing ERG electrodes and recording the ERG. It is of

monitor paramount importance that the input current of the amplifier used to record the ERG be extremely low as series damage can occur to the eye if DC current is allowed to flow through an ERG electrode. Since stimulus artifact is not involved, AC-coupled amplifiers can be used and a capacitor may be used in series with the ERG electrode to eliminate DC currents.

caution

components
of ERG

A typical ERG is shown in Fig. 11-13 together with a diagram showing the general characteristics of the ERG. Variations in the characteristics of the stimulating light, as well as variations in the level of the state of light adaptation of the retina, will affect the response characteristics. The electroretinogram shown in Fig. 11-13 was recorded after the subject had become adapted to room lighting. The general characteristics of the electroretinogram consist of an initial A wave, followed by a positive B wave, a more slowly developing positive C wave (which is dependent upon the duration of the stimulation) and a D wave or off-effect.

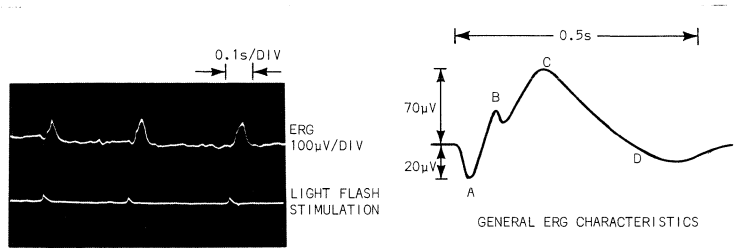


Fig. 11-13. A typical ERG.

12

STIMULATION - ELECTROMYOGRAPHY
- NERVE CONDUCTION

12.1 STIMULATION

Chapter 1 covers the bioelectric potentials generated within individual cells when these cells are stimulated. Stimulation refers to an external force being applied to the cell which results in the cell depolarizing and then repolarizing. Cells in the eye are sensitive to light stimulation, cells in the ear are sensitive to sound stimulation and nerve cells are sensitive to electro-chemical stimulation. All cells are sensitive to one degree or another to artificial electrical stimulation: The passage of electric current through the cell will cause it to depolarize.

artificial
stimulation

When analyzing biomedical phenomena it is often desirable to artificially stimulate a group of cells. Such artificial stimulation is achieved by using a pulse generator to pass current through the cells concerned for a brief period of time. If, for example, one wishes to cause the ulnar nerve in the arm to propagate a depolarization pulse, this nerve can be stimulated by passing current through the arm with the aid of two surface electrodes placed so that part of the current flow between them passes through the ulnar nerve.

In general, it is impossible to localize stimulating current to a particular cell or small group of cells. Current is passed through the bulk tissue surrounding these cells and a finite, *but unpredictable*, portion of this current will pass through the cells. Since it is *impossible* to *localize* current flow to a particular cell, it is impossible to know how much current will flow through it.

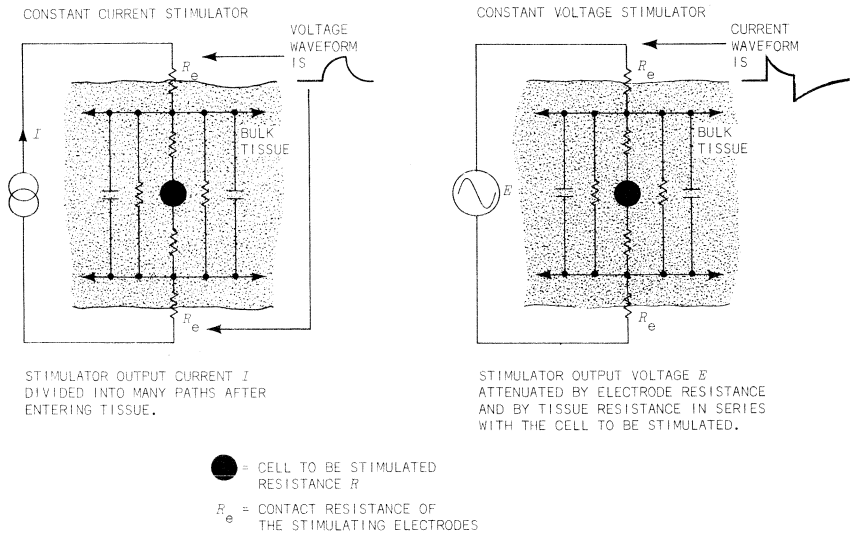


Fig. 12-1. Constant current and constant voltage stimulation.

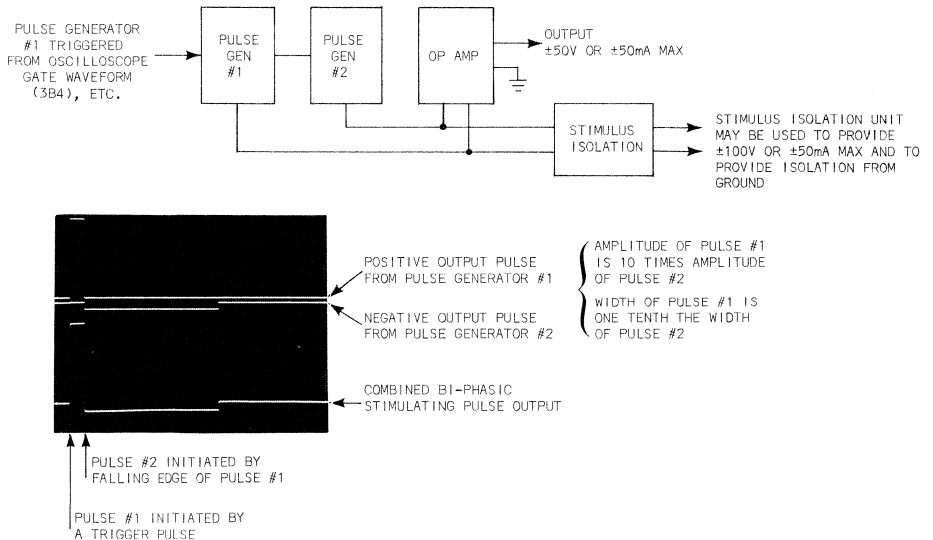


Fig. 12-2. Biphasic stimulation pulse generator.

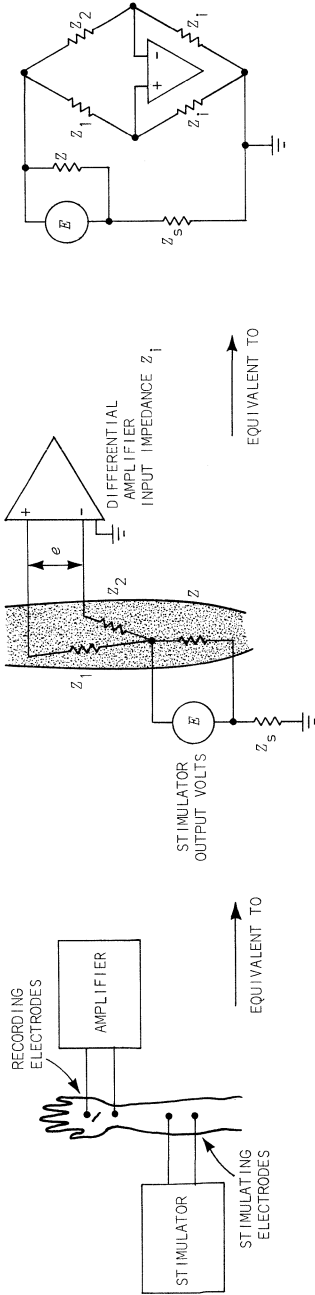
stimulator
types

Stimulators provide an output from either a constant current source ($Z_0 > 10,000$ ohms) or from a constant voltage source ($Z_0 < 50$ ohms). Both constant current stimulators and constant voltage stimulators are extensively used and arguments for and against constant current or constant voltage stimulation have existed since the very early days of physiology. Although this controversy as to the best type of stimulator for routine use still exists, neither type allows an accurate prediction of the amount of current passing through the cells concerned. Satisfactory results can be obtained with either type. Many modern pulse generators intended for tissue stimulation provide both constant current and constant voltage output characteristics. A pair of stimulating electrodes immersed in tissue fluid typically has an impedance of about 500 ohms at the frequencies involved in the stimulus pulse.

depolarization

If a stimulus current of insufficient intensity to cause cell depolarization is applied to a cell, the cell membrane resting potential will be reduced from its normal -90 millivolts. If an additional stimulus pulse of the same intensity is then applied to the cell within the next few milliseconds or so, the cell may then depolarize, as the cell membrane potential has not yet returned to its normal resting potential, as discussed in Chapter 1. The cell would thus appear to be more sensitive to stimulation from the second pulse than from the first. It has been suggested that cell recovery from the effect of a stimulus current can be hastened by the passage of an opposing, lower intensity, current for an appreciably longer period so that the net quantity of electricity is zero. This feature is incorporated in the biphasic stimulator shown in Fig. 12-2. Biphasic stimulation also neutralizes recording electrode polarization where silver/silver chloride recording electrodes are not used and helps to maintain a fixed baseline on DC-coupled recording systems. In the waveform shown in Fig. 12-2, the stimulating pulse is followed by a pulse of opposite polarity, of one-tenth the amplitude and ten times the width.

biphasic
stimulation



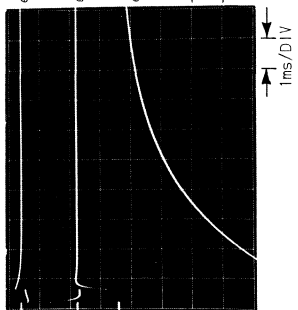
- Z_s = STIMULATOR IMPEDANCE TO GROUND (ZERO FOR GROUNDED STIMULATOR)
- Z = IMPEDANCE LEAD TO STIMULATOR AND AMPLIFIER + INPUT
- Z_1 = TISSUE IMPEDANCE BETWEEN STIMULATOR AND AMPLIFIER + INPUT
- Z_2 = TISSUE IMPEDANCE BETWEEN STIMULATOR AND AMPLIFIER - INPUT
- Z_i = INPUT IMPEDANCE OF AMPLIFIER

E = OUTPUT VOLTAGE OF STIMULATOR
 e = DIFFERENTIAL SIGNAL BETWEEN AMPLIFIER + AND - INPUTS

$$e \propto \frac{(Z_2 - Z_1)}{Z_s + Z_1}$$

SINCE $Z_2 - Z_1$ WILL NEVER BE ZERO
 e CAN BE MINIMIZED BY MAKING Z_s AND Z_1 LARGE:
 TYPICAL $Z_s = 20\text{pF}$, $10^9\Omega$ AND $Z_1 = 30\text{pF}$, $10^7\Omega$

e PRODUCED WITH 100V STIMULATION AT 1mV/DIV RECORDED 5cm FROM STIMULATOR



e STIMULUS ARTIFACT WITH GROUNDED STIMULATOR AND 100mΩ, 2pF AMPLIFIER
 e STIMULUS ARTIFACT WITH ISOLATED STIMULATOR AND 1MΩ, 47pF AMPLIFIER
 e STIMULUS ARTIFACT WITH GROUNDED STIMULATOR AND 1MΩ, 47pF AMPLIFIER

Fig. 12-3. Equivalent circuit of stimulator, subject and amplifier.

12.2 STIMULUS ISOLATION

The previous section referred to stimulation. Stimulation may be produced directly from a pulse generator or it may be produced via a stimulus isolation unit. The results of stimulation may be detected with an oscilloscope and it is important that the pulse produced by the stimulator does not interfere with the response obtained on the oscilloscope. It is thus necessary to insure that the stimulating current is limited to the area between the two stimulating electrodes and that little or no stimulating current appears in the region of the recording electrodes.

Any stimulating current flowing near the recording electrodes will cause a potential difference between these electrodes which will appear as an out-of-phase signal and thus not be rejected by the differential amplifier. The equivalent circuit involved in stimulating and recording is shown in Fig. 12-3. In this circuit, most of the stimulating current flows through the impedance formed by the tissue between the stimulating electrodes. It is apparent, however, that two alternative current paths exist; via the tissue impedances Z_1 and Z_2 and the amplifier input impedances to ground. If it were possible to make Z_1 equal to Z_2 , then no in-phase signal would be produced at the recording electrodes. These impedances are not controllable however, and it is impossible in practice to balance them by careful placement of the recording electrodes.

reducing
stimulus
artifact

Since it is not possible to balance the tissue impedances, then one must reduce the stimulating current passing through these impedances to as low a value as possible in an effort to reduce the differential signal appearing at the recording electrodes. To reduce these currents, it is necessary to either increase the input impedance of the differential amplifier or to increase the impedance between stimulator and ground. In practice one attempts to make both these impedances as large as possible in an effort to achieve the maximum reduction in the level of stimulus signal appearing at the recording electrodes. With a nonisolated stimulator, the impedance between stimulator and ground is zero, however with an isolated stimulator this impedance can be made very high, typically 20 picofarads and 10^9 ohms. A typical differential amplifier input impedance would be 30 picofarads and 10⁷ ohms

isolated
stimulator

high input
Z amplifier

If a grounded stimulator is used on tissue, the ground electrode should *always* be placed between the "hot" stimulating electrode and the recording electrodes, to minimize stimulus artifacts. The upper trace shown on the waveform in Fig. 12-3 shows an acceptable level of stimulator pulse appearing at the recording electrodes when the recording electrodes are approximately 5 centimeters from the stimulating electrodes. This trace shows a 100 volt stimulator pulse reduced to less than 0.5 millivolts, a reduction of 200,000:1, which is entirely acceptable. The second trace shown on the waveform in Fig. 12-3 is also acceptable as, although the level of stimulator pulse appearing at the recording electrodes is substantially greater than shown on the upper trace, the amplifier quickly returns to zero after the stimulating pulse returns to zero. The lower trace shown on the waveform in Fig. 12-3 is unacceptable. In this instance, the stimulating pulse has severely overloaded the differential amplifier and the amplifier takes longer than 10 milliseconds to return to zero. Any response occurring in this 10 millisecond period would be either camouflaged by, or distorted by, this amplifier overload characteristic.

acceptable
artifacts

amplifier
overload

In practice, complete reduction of the stimulator pulse is rarely achieved and a severely attenuated stimulator pulse will appear on the oscilloscope trace. This attenuated stimulator pulse is referred to as stimulus artifact. Stimulus artifact is not altogether undesirable if it is not of excessive amplitude as it does provide a time reference on the CRT trace.

The above discussion assumes the use of DC-coupled differential amplifiers. If AC-coupled differential amplifiers are used, the time constant of the input coupling capacitors and the input resistance will determine the amplifiers recovery characteristics after overload.

AC-coupled
amplifiers

12.3 STRENGTH/DURATION CURVES

Strength/duration curves show the excitability characteristics of muscle or nerve fibers. Stimulation of a cell is achieved by the passage of a certain *quantity* of electricity through the cell,

thus stimulation is dependent on charge rather than on current. Stimulation can be achieved by passing a large current for a short period or with a lesser current for a longer period. Since diffusion within the cell tends to oppose the stimulating current, a lower limit of current is reached below which stimulation will not occur, no matter how long the current is maintained.

For any given muscle or nerve fiber a strength/duration curve may be drawn, as shown in Fig. 12-4, which represents the minimum stimulus or threshold required to stimulate the muscle or nerve fiber. Referring to the strength/duration curve shown in Fig. 12-4 for a normal muscle, it can be seen that a 0.01 millisecond stimulus pulse of 3.5 amplitude units applied in the region of the nerve associated with the muscle will cause the nerve to depolarize, causing muscular action. Alternatively, muscle action can be produced with a 0.03 millisecond wide stimulus pulse of 1.7 amplitude units, a 0.1 millisecond wide stimulus pulse of 1.1 amplitude units or a 0.3 millisecond wide stimulus pulse of 1.0 amplitude units. It is also apparent that, for pulse widths greater than 0.3 millisecond, no further reduction in stimulus amplitude will be effective. This amplitude, below which stimulation will not occur, no matter what the stimulus pulse width is, is referred to as the nerve tissue threshold or *rheobase* for the muscle. The term is now rarely used.

nerve tissue
threshold

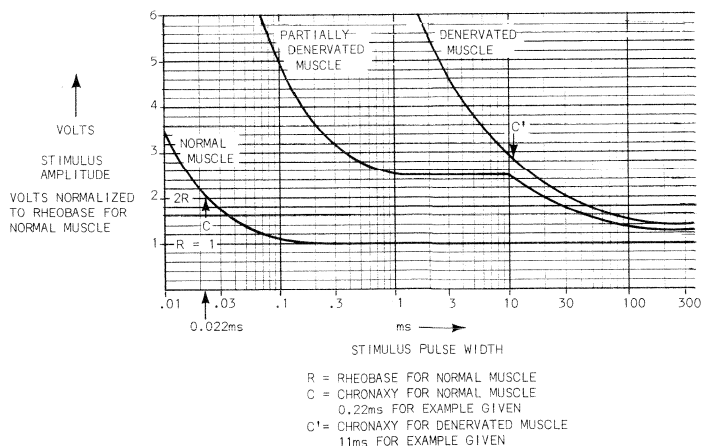


Fig. 12-4. The strength/duration or S-D curve.

Pulse widths of from 0.03 millisecond to 100 millisecond in eight steps have been proposed as an international standard for strength/duration curve determination.

In the particular example shown in Fig. 12-4 this rheobase occurred when using a constant voltage stimulator set at 30 V. Since, as discussed earlier, this absolute voltage level has little meaning, strength/duration curves are normally shown with a stimulus amplitude normalized to the rheobase, that is 30 volts in this instance is equivalent to 1.0 unit of stimulus amplitude, 2.0 units of stimulus amplitude being equivalent to 60 volts, etc.

pulse
amplitude
normalized
to rheobase

Once a strength/duration curve has been determined, it is desirable to have a technique whereby curves for various muscles can be compared. Strength/duration curves may be compared by comparing the chronaxy obtained from the curve. The chronaxy is the minimum pulse width required to excite the tissue for a stimulus of twice the amplitude of the rheobase. The chronaxy, in the example shown on Fig. 12-4, is 0.022 millisecond for normal muscle.

chronaxy

denervated
muscle
action

The strength/duration curve for a denervated muscle is also shown in Fig. 12-4. A denervated muscle exists when the nerve connection to the muscle has effectively been interrupted. This may be achieved by impairing the motor end plate action with drugs such as curare. It can be seen from Fig. 12-4 that a denervated muscle is less sensitive to stimulus than a normal muscle. In a normal muscle the stimulating current excites the more sensitive nerve fibers which in turn excites the muscle whereas in a denervated muscle the stimulating current must stimulate the muscle directly. Thus the strength/duration curve for a normal muscle shows the excitability characteristics for a single motor neuron, however, the strength/duration curve for a denervated muscle shows the excitability characteristics of a single muscle fiber.

Fig. 12-4 also shows a strength/duration curve for a partially denervated muscle. It can be seen, for pulse widths up to 10 milliseconds, muscle action is due to stimulation of the nerve whereas for pulse widths greater than 10 milliseconds muscle action is due to stimulation of the muscle directly. The strength/duration curve is, thus, displaying the excitability characteristics of the component which has the lower threshold for a specific pulse width.

muscle
reaction to
stimuli

When preparing strength/duration curves, muscle action is determined by viewing the muscle concerned in a well-lit environment. When stimulating with a 100 millisecond pulse, the muscular contraction produced is brisk and pronounced when the contraction is due to excitation of the nerve fibers and is sluggish and wormlike when the contraction is due to direct excitation of the muscle fibers. When stimulating with narrower pulses, muscle action is characterized by a twitching of the muscle concerned. Muscle action can also be determined by using some form of force-gage attached to the muscle concerned to detect muscular movement or, more commonly, by recording the electrical activity produced within the muscle when the muscle is stimulated by the nerve.

abbreviated
strength/
duration
curves

In some instances, the plotting of the strength/duration curve is somewhat tedious and abbreviated strength/duration curves are obtained by determining the stimulus amplitude required for a 100 millisecond pulse and a 1 millisecond pulse and expressing these as a ratio. In a normal muscle this ratio should be approximately unity, in a partly denervated muscle it will be between 1.5 and 4 and in a fully denervated muscle it will be between 4 and infinity. Abbreviated strength/duration curves may also be obtained by determining the rheobase and then determining the chronaxy directly by doubling the stimulus amplitude and reducing the pulse width to a point where the tissue can no longer be stimulated. This point will represent the chronaxy.

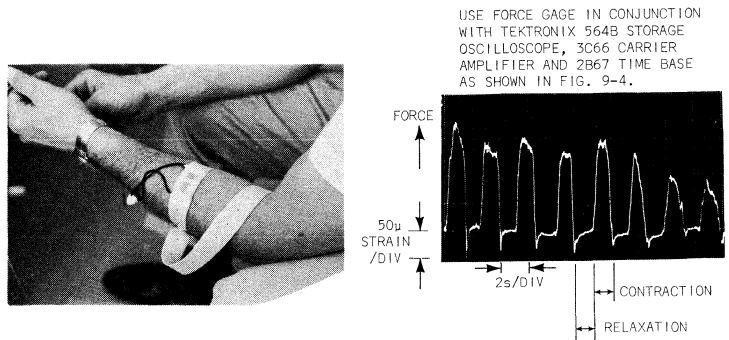
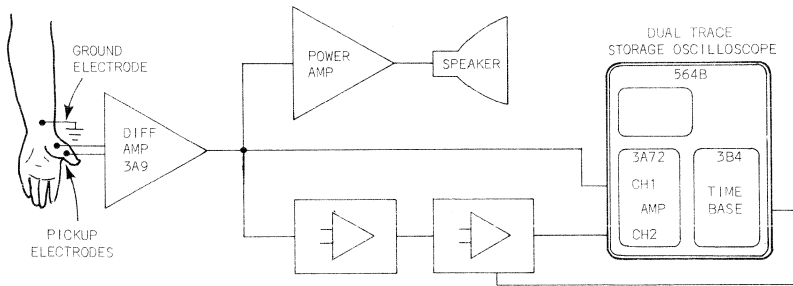


Fig. 12-5. A myograph using an elastic force gage.

12.4 MYOGRAPHY

strain
gage

Myography is the study of muscular contractions and a myograph is an apparatus for recording the mechanical effects of a muscular contraction. A myograph may simply consist of a displacement transducer or a force transducer mechanically coupled to the muscle under investigation. As shown in Fig. 12-5, an elastic strip is placed around the muscle concerned and a strain gage is bonded to this elastic strip. Muscular contraction causes a tension increase in the elastic strip which results in a resistance change in the strain gage. The muscular contraction may be initiated voluntarily or produced by electrical stimulation. A strain gage myograph may be used with a Tektronix Type 564B Storage Oscilloscope and a Tektronix Type 3C66 Carrier Amplifier as shown in Chapter 9, Fig. 9-4. The output from such a recording system, a series of muscular contractions over a 20-second period, is also shown in Fig. 12-5. Force myographs are particularly suited to exercising subjects or for the study of muscular fatigue over prolonged periods.



3A9 AND 3A8 USED IN ANY
TEKTRONIX 560 SERIES
OSCILLOSCOPE (e.g., 561B).

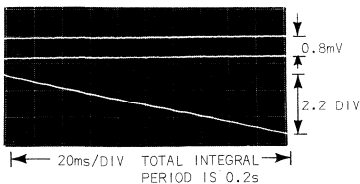
ABSOLUTE VALUE
CIRCUIT - "FULL
WAVE RECTIFIER"

INTEGRATOR GATED
WITH TIME BASE

3A8 OPERATIONAL AMPLIFIER
WITH ABSOLUTE VALUE ADAPTER
AND GATING ADAPTER.
(SEE CHAPTER 29 FOR DETAILS)

- 3A9 DIFFERENTIAL AMPLIFIER SET AT 1mV/DIV. PROVIDES A GAIN OF 1000X TO THE 3A8 OPERATIONAL AMPLIFIER AND 3A72 CHANNEL 1.
- 3A72 CHANNEL 1 SET AT 1V/DIV PROVIDING A SENSITIVITY REFERRED TO THE ELECTRODES OF 1mV/DIV.
- 3A8 OPERATIONAL AMPLIFIER #1 USED IN CONJUNCTION WITH AN ABSOLUTE VALUE CIRCUIT AS SHOWN IN CHAPTER 29.
- 3A8 OPERATIONAL AMPLIFIER #2 USED IN CONJUNCTION WITH A LOW SPEED GATING ADAPTER AS SHOWN IN CHAPTER 29.
 Z_1 SET AT 1m Ω , Z_2 AT 0.1 μ F FOR INTEGRATION.
- 3A72 CHANNEL 2 SET AT 0.1, 0.2 OR 0.5V/cm PROVIDING VARYING INTEGRATION SENSITIVITIES.

INTEGRATOR CALIBRATION - OSCILLOSCOPE AMPLIFIER #2 AT 0.2V/DIV



4mV CALIBRATOR SIGNAL FROM OSCILLOSCOPE CALIBRATOR CONNECTED TO DIFFERENTIAL AMPLIFIER.

DIFFERENTIAL AMPLIFIER GAIN REDUCED BY 5X (FROM 1mV/DIV TO 5mV/DIV) PROVIDING A CALIBRATION VOLTAGE RELATIVE TO THE EMG SYSTEM OF 0.8mV.

CALIBRATOR DUTY FACTOR IS 50%.

$$\begin{aligned} \text{CALIBRATOR INPUT SHOWN} &= 0.8 \times 10^{-3} \times \frac{50}{100} \times 0.2 \text{ VOLT SECONDS} \\ &= 80 \times 10^{-6} \text{ VOLT SECONDS.} \end{aligned}$$

2.2 DIV OF INTEGRATOR OUTPUT = 80×10^{-6} VOLT SECONDS
 \therefore INTEGRATOR CALIBRATION IS 36×10^{-6} VOLT SECONDS/DIV WITH OSCILLOSCOPE AMPLIFIER #2 AT 0.2V/DIV.

$$\begin{aligned} \text{INTEGRATOR OUTPUT} &= \\ &180 \times 10^{-6} \text{ VOLT SECONDS INPUT/VOLTS OUTPUT} \end{aligned}$$

Fig. 12-6. An electromyograph system.

12.5 ELECTROMYOGRAPHY -- EMG

muscle
fiber
depolarization

Whereas the myograph records the mechanical effects of a muscular contraction, the electromyograph records the electrical effects of such a contraction. Muscular contraction is caused by depolarization of the muscle fibers. This depolarization produces an action potential as covered in Chapters 1 and 3. This muscular action potential is known as the electromyogram, or EMG. An electromyogram will be produced in a muscle when the muscle contraction is caused either by voluntary muscle action or by electrical stimulation of the muscle.

12.6 ELECTROMYOGRAPHY WITH VOLUNTARY MUSCULAR ACTION

instrumentation

A typical system for recording the electromyograph produced by voluntary muscle action is shown in Fig. 12-6. Further details on this system are given in Chapter 29. The muscle action potential is picked up by needle electrodes inserted into the muscle or by surface electrodes placed over the muscle concerned and then amplified by a suitable differential amplifier. The EMG can then be detected audibly by using a speaker in conjunction with an audio amplifier. The EMG may also be displayed directly on an oscilloscope or may be converted to an absolute integral and then displayed on an oscilloscope. Both direct displays and integrated displays are shown in Fig. 12-7.

absolute integrals

Referring to Fig. 12-7, the upper trace on the top photograph shows an EMG produced by a mild voluntary contraction. The action potential produced by a single motor unit can clearly be differentiated from other action potentials. The absolute integral of this activity is displayed on the lower trace in the same photograph. The integral displays the quantity of electricity involved in the muscular contraction. The quantity of electricity associated with the single motor unit action potential can also be determined.

With a more forceful voluntary contraction, as shown in the middle photograph, many motor units are involved and the EMG obtained is the result of the action potential produced by all these motor units; the resulting integral being greater than the integral obtained for a single motor unit.

By using a slower sweep speed in conjunction with the system shown in Fig. 12-6, the EMG and absolute integral for a series of contractions may be displayed as shown in the bottom photograph.

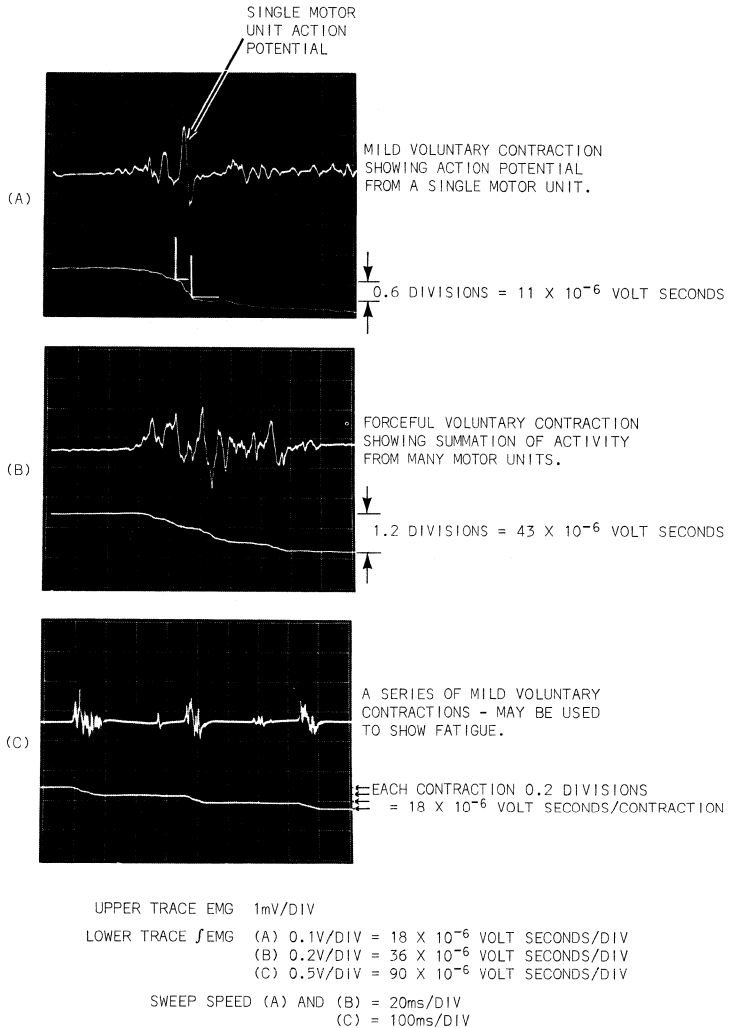


Fig. 12-7. Results obtained with the system shown in Fig. 12-6.

electrical
quantity

volt
seconds

The quantity of electrical activity produced by a muscular contraction of a montomic function of the strength of the contraction. Since it is difficult to estimate this quantity from an observation of the EMG waveform, the absolute integral of the EMG is used as a measure of this quantity. The integrator output is calibrated in units of electrical quantity, that is, volt seconds. Since one volt second is a substantial quantity of electricity, it is preferable to refer to EMG's in smaller units, one "unit" being equal to 10^{-6} volt seconds. It can be seen from Fig. 12-7 that for a particular subject and a particular muscle 11 units are produced by a single motor unit, 18 units by a mild muscular contraction and 43 units by a forceful contraction.

calibration

Referring to Fig. 12-6, the absolute integral is obtained by full-wave rectification of the electromyograph and then integrating this rectified signal with an integrator gated "on" for the duration of the oscilloscope sweep. The whole system consisting of the differential amplifier, the full-wave rectifier, the integrator and the display oscilloscope can be calibrated using an oscilloscope calibrator waveform as shown in Fig. 12-6. The system shown in Fig. 12-6 may be duplicated by using one Type 561B Oscilloscope in conjunction with a Type 3A9 Differential Amplifier and a 3A8 Operational Amplifier and one Type 564B Storage Oscilloscope in conjunction with a Type 3A72 Dual Trace Amplifier and a 3B4 Time Base unit. Details of the absolute value adapter and gating adapter for the 3A8 are given in Chapter 29.

audio
output

The EMG is often presented audibly in clinical applications and the trained listener can judge the condition of the muscle by the volume and characteristic tones produced by the audio system during a muscular contraction.

12.7 ELECTROMYOGRAPHY DURING ELECTRICAL STIMULATION

voluntary
versus
stimulated
EMG

The EMG produced during a voluntary twitch is spread out over a period of 100 milliseconds or more as the nerve impulses to various motor units are not time coincident as the propagation delay from the spinal cord to the muscle concerned is different for all nerve fibers. Also, since the contraction is voluntary, any one motor unit may produce several action potentials, the frequency of discharge being determined within the spinal cord. Such is not the case when recording EMG produced by electrical stimulation. All neurons with thresholds above the stimulating intensity are simultaneously stimulated by the electrical impulse, thus all muscle fibers discharge simultaneously, producing substantial activity for a brief period of time, typically less than 10 milliseconds. Although the response obtained when stimulating is referred to as an EMG, it is an unnatural occurrence and should perhaps be more correctly referred to as a "myographic response" or a "muscle action potential." The stimulus pulse used to initiate this response usually has an amplitude of >100 volts and is either 0.1 millisecond, 0.3 millisecond, or, occasionally, 0.5 millisecond wide.

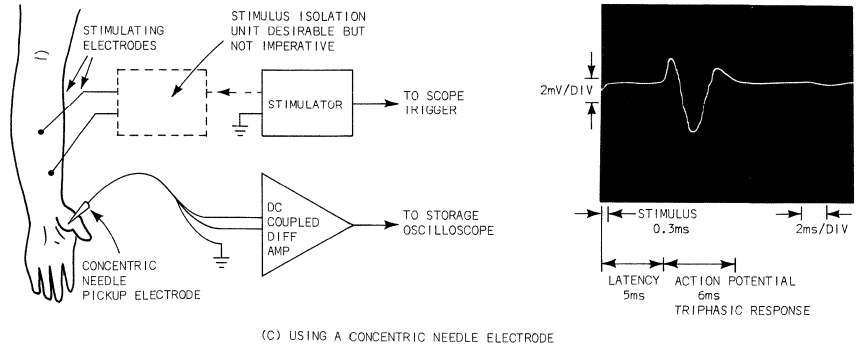
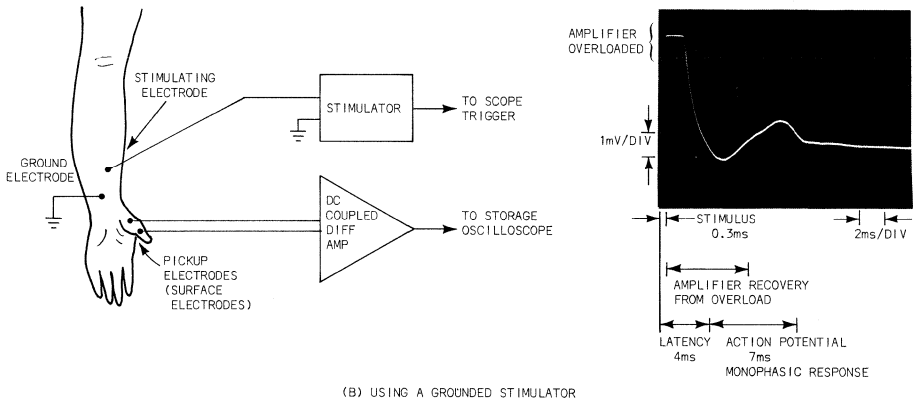
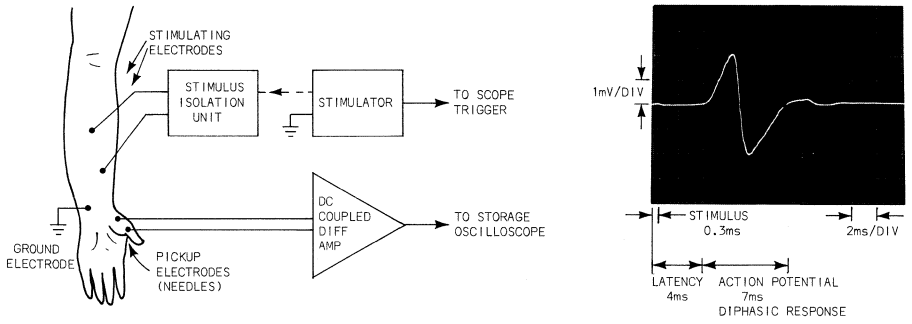


Fig. 12-8. EMG produced by electrical stimulation.

A typical EMG produced by electrical stimulation is shown in Fig. 12-8A. A 0.3 millisecond stimulus pulse was used to initiate the response. A delay occurred between the stimulating pulse and the response; this delay is referred to as latency. The action potential shown in Fig. 12-8A has a latency of 4 milliseconds and produces an action potential covering the following 7 milliseconds. This action potential is referred to as diphasic as it shows a single positive deflection followed by a single negative deflection.

latency

diphasic
response

The action potential shown in Fig. 12-8B is referred to as monophasic as the action potential appears to be comprised of only one positive deflection. This action potential is partially camouflaged by the amplifier recovery characteristics due to the use of a nonisolated stimulator. Such amplifier characteristics were covered earlier in this chapter. If a grounded stimulator is used on tissue, the ground electrode should *always* be placed between the "hot" stimulating electrode and the recording electrodes, to minimize stimulus artifacts.

monophasic
response

The EMG shown in Fig. 12-8C is referred to as triphasic as two positive deflections and one negative deflection are exhibited. This EMG was recorded using a concentric needle pickup electrode which effectively locates two electrodes less than 1 millimeter apart in the muscle in an attempt to record the action potential produced by a single motor unit rather than by the complete muscle. All three EMG's shown in Fig. 12-8 are considered acceptable and basically serve to show that the nerve and muscle relationship is functioning correctly as the latency is not excessive and an action potential is, in fact, generated.

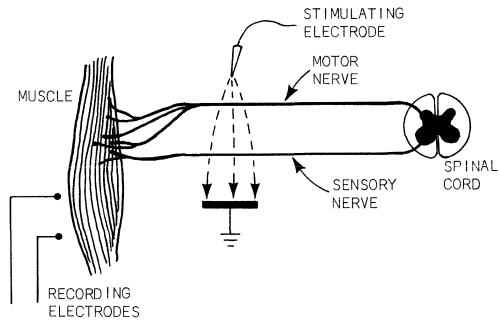
triphasic
response

12.8 THE H REFLEX

The muscular reflex response generated within the spinal cord was covered in Chapter 3. When recording EMG's produced by electrical stimulation, the stimulating current excites the motor nerve which in turn initiates a response in the muscle concerned.

muscle
response
via
reflex
stimulation

This stimulating current also excites the sensory nerve. It is possible to decrease the stimulus level to a point where the stimulus intensity is insufficient to excite the motor nerve but is sufficient to excite the more sensitive sensory nerve. The depolarization pulse propagated in the sensory nerve as a result of this stimulation travels to the spinal cord where a reflex response is in turn propagated in the motor nerve. This reflex response propagates along the motor nerve to the muscle concerned, initiating a muscular response.



MUSCLE ACTION PRODUCED BY STIMULUS CURRENT DEPOLARIZING MOTOR NERVE (DIRECT) OR BY STIMULUS CURRENT DEPOLARIZING SENSORY NERVE WHICH IN TURN DEPOLARIZES MOTOR NERVE DUE TO REFLEX RESPONSE GENERATED IN SPINAL CORD (INDIRECT).

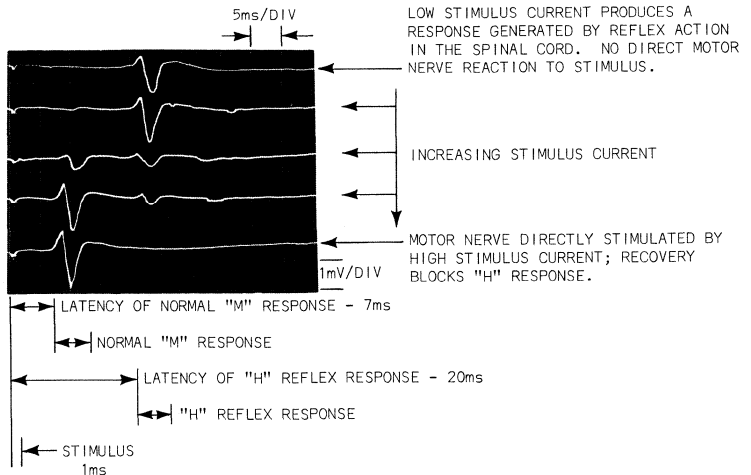


Fig. 12-9. The H-Reflex response.

H reflex

The physical relationship between the muscle, the nerve and the spinal cord is shown in Fig. 12-9, together with the results obtained when recording the EMG with progressively increasing stimulus intensity. At low stimulus intensity a response having a latency of approximately 20 milliseconds is detected, known as the "H reflex" response. This latency is due to the conduction time from the stimulating point along the sensory nerve to the spinal cord and thence from the spinal cord along the motor nerve to the muscle concerned. As the stimulating current is progressively increased this H reflex response decreases and a normal or "M reflex" response appears with a normal latency of 7 milliseconds, representing the conduction time from the stimulus site, via the motor nerve, to the muscle concerned. The "H reflex" response can be used to determine the condition of the reflex system.

M reflex

12.9 NERVE CONDUCTION

motor nerve
propagation
velocity

The propagation velocity of the nerve impulse along the motor nerve from the stimulus site to the muscle can be determined as shown in Fig. 12-10. In the example shown, the peroneal nerve of the left leg is stimulated behind the knee and a muscular response is detected in the foot, using either surface electrodes or needle electrodes. The response shown has a latency of 11.5 milliseconds. The stimulus electrodes are then moved to a point behind the ankle and a response obtained in the foot having a latency of 4 milliseconds. The difference between these two latencies is attributed to the conduction time required for the nerve impulse to propagate along the motor nerve from the knee to the ankle.

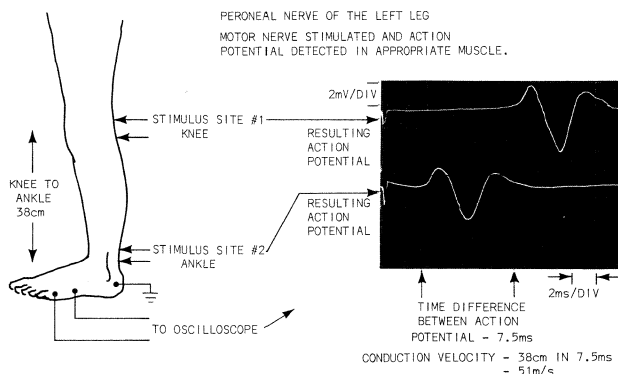


Fig. 12-10. Motor nerve conduction velocity determination.

The propagation velocity can be determined by measuring the distance from the knee stimulation point to the ankle stimulation point and dividing it by the difference in latencies. A 100 volt stimulus pulse, 0.3 millisecond wide, is usually used. Occasionally, pulse widths of either 0.1 millisecond or 0.5 millisecond are used. The stimulus should be repeated several times to ensure that the response obtained is consistent.

sensory
nerve
propagation
velocity

A similar technique can be used to measure sensory nerve conduction velocity as shown in Fig. 12-11. In Fig. 12-10 we moved the stimulus site with respect to a fixed recording site, however, when attempting to record sensory nerve conduction velocity, it is necessary to stimulate at a fixed sense receptor site and to record the propagation of this stimulus pulse along the sensory nerve by detecting the nerve impulse or "traveling wave of depolarization" at various sites along the nerve. Fig. 12-11 shows the results obtained when stimulating the hand and recording the propagation of this pulse along the ulnar nerve at four points along the length of the nerve using a four-channel oscilloscope. If the vertical position of the four channels on this oscilloscope are adjusted to represent distance in centimeters from the stimulus site, the various latencies obtained should be directly proportional to this distance, resulting in the straight line shown in Fig. 12-11. Any deviation in straightness in this line would represent a change in conduction velocity. Injury to the nerve will normally result in decreased conduction velocity in the injured part of the nerve.

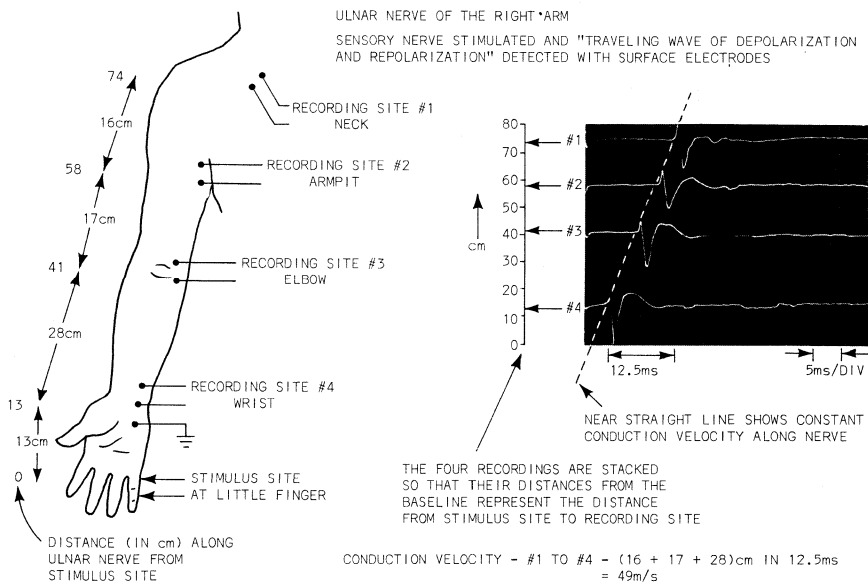


Fig. 12-11. Sensory nerve conduction velocity determination.

limb nerves investigated When measuring conduction velocity four nerves are principally investigated: the ulnar and median nerves of the arm and the peroneal and tibial nerves of the legs. The ulnar, peroneal and tibial nerves perform both general sensory and motor functions, however the median nerve primarily performs a sensory function. Responses from the ulnar nerve are normally detected on the back of the hand or on the fingers, the median nerve on the thumb or on the thick fleshy part of the hand near the thumb, the peroneal nerve on the four smaller toes or the top of the foot and the tibial nerve at the side of the foot near the larger toe or on the larger toe.

40 to 60 m/s normal Normally, conduction velocity is measured, or EMG's are recorded, in an attempt to diagnose an abnormality in the subject and, since this abnormality usually manifests itself in only one side of the subject at a time, the limb on the other side of the subject may be regarded as normal and conduction velocities or EMG's compared between the two limbs. Although conduction velocity depends on the nerve under investigation, conduction velocities in most healthy nerves fall in the range from 40 to 60 meters per second; injury or abnormality being indicated by lower conduction velocities extending to below 10 meters per second. The term electrodiagnosis is applied to the general study of normal motor unit behavior. Complete electrodiagnosis should include most of the techniques covered in this chapter.

12.10 REPETITIVE STIMULATION

recovery characteristics So far this chapter has dealt with the effects of a single stimulating pulse on muscle and nerve fiber. In attempting to determine the recovery characteristics of a motor unit it is necessary to stimulate with a double pulse and to determine the delay required between the two pulses for the stimulus pulse to be "seen" by the muscle as two separate stimuli rather than as only one stimulus. This minimum pulse separation, when translated to frequency, is known as the critical frequency. The critical frequency varies for different muscles in the same subject and for the same muscle at a different temperature and in a different state of fatigue. The critical frequency for most of the major muscles in the human body lies between 5 and 15 Hz corresponding to minimum pulse separations of 200 milliseconds and 66 milliseconds.

critical frequency

Multiple pulse stimulation, rather than single pulse or double pulse stimulation, is used to determine the fatigue characteristics of nerve and muscle fiber. Under normal conditions, the response obtained during prolonged stimulation should show little change from the response obtained from a single stimulus as long as a brief relaxation period is allowed between each pulse.

12.11 SMOOTH MUSCLE POTENTIALS

Previously in this chapter we have dealt with the electrical activity produced in muscles known as skeletal muscle. Skeletal muscle produces contractions in a series of muscle fibers, the combined effect producing continuous motion. Other muscles, particularly the muscles surrounding the major organs in the torso, produce overall continuous contraction and are known as "smooth" muscles. "Smooth" refers to the microscopic appearance of the muscle, in contrast to "striated" skeletal muscle, with its characteristic cross stripes. Electrodes placed on, in, or over smooth muscle can be employed to detect contractions in these muscles. Electrodes appropriately inserted into the stomach, bladder, etc. or placed on the surface of the body over these organs can serve to monitor the slowly varying potentials generated by the muscles in these organs. These potentials are known as smooth muscle potentials. Although these potentials are rarely monitored, the recording of the electrical activity produced by the stomach has been termed the electrogastrogram. The electrogastrogram is characterized by a slowly changing "DC" potential (below 1 Hz) which would normally be recorded on a DC-coupled instrument at a sensitivity of 10 millivolts per division and at a sweep speed of perhaps 1 second per division.

electro-
gastrogram

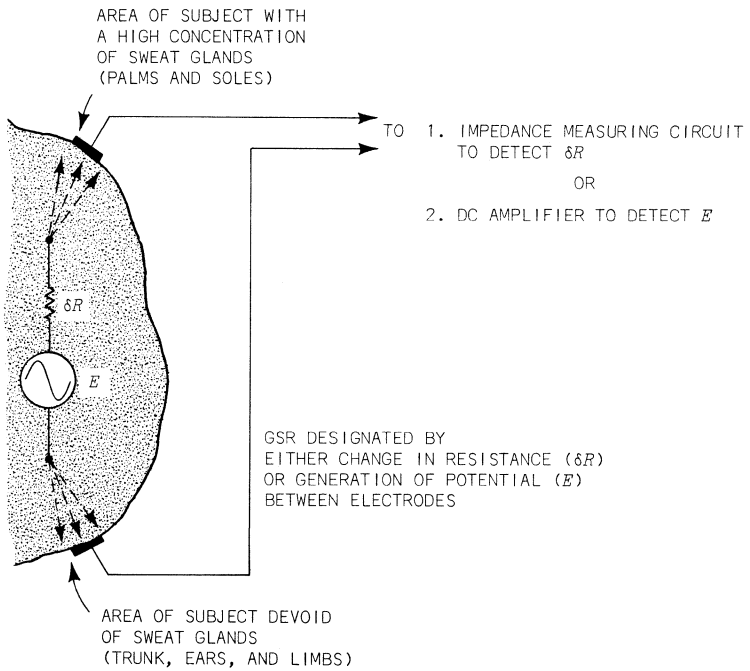


Fig. 13-1. GSR measurement.

13

GALVANIC SKIN REFLEX - GSR

13.1 THE AUTONOMIC NERVOUS SYSTEM

The human autonomic nervous system is the system within the body that regulates body functions such as temperature, respiration and glandular activity. When a subject is psychologically excited or is in some other elevated state of psychological activity, the subject perspires or "sweats." This is due to an emotional stimulus initiating a response in the autonomic nervous system which in turn produces a response in the subject's sweat glands. Detection of this sweat gland activity is, thus, an indication of the subject's psychological state or state of arousal.

sweat
gland
activity

Activity of the sweat glands is referred to by one or more of the following terms: electrical skin resistance (ESR), galvanic skin reflex (GSR), electrodermal response (EDR) and psychogalvanic reflex (PGR). Occasionally, the term GSR is referred to as the galvanic skin reaction or galvanic skin resistance. These terms all relate to one or both of the following physiological changes associated with sweat gland activity: a change in resistance and the generation of a potential between areas containing many sweat glands and areas almost devoid of them. The change in resistance is referred to as the Fere effect on the exosomatic response of the GSR. A decrease in the subject's resistance indicates arousal. Relaxation is indicated by an increase in resistance. The generation of a potential difference is referred to as the Tarchanoff effect or the endosomatic response of the GSR. This resistance change and potential generation is represented by δR and E in Fig. 13-1.

Fere
effect

Tarchanoff
effect

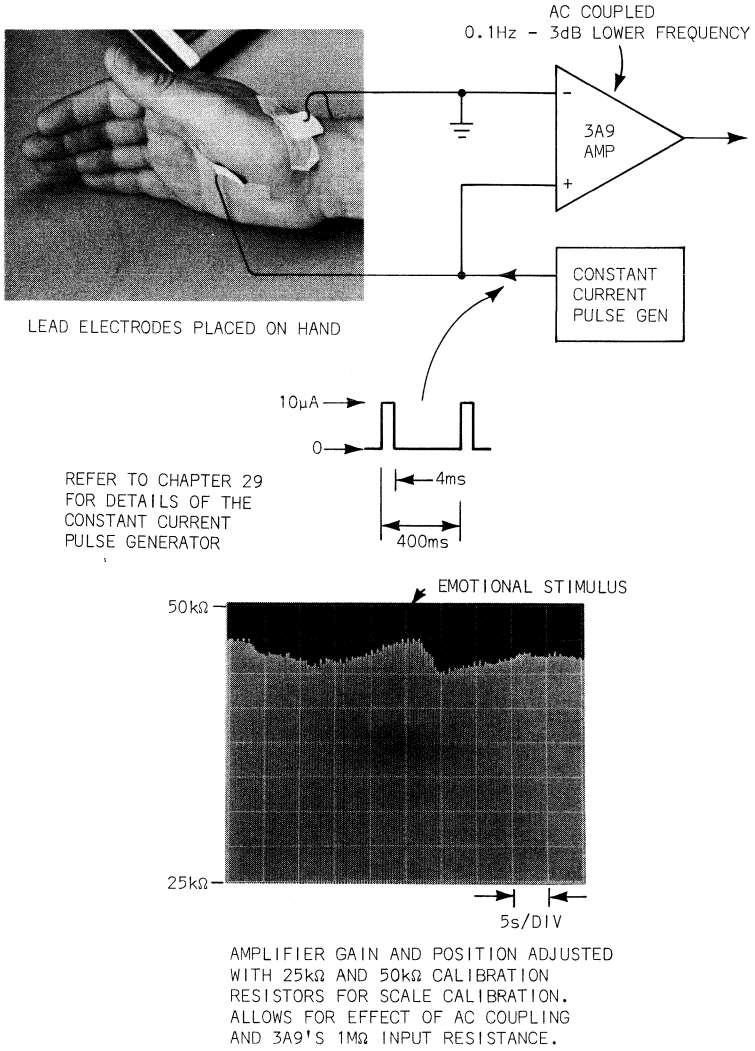


Fig. 13-2. GSR measurement.

13.2 GSR MEASUREMENT BY RESISTANCE CHANGE

DC
techniques

GSR measurement by this technique involves the detection of an impedance change between two electrodes on the subject. The GSR primarily changes the resistive component of this impedance, it is thus important that the measurement technique used be insensitive to reactive component changes. The simplest technique would appear to be the passage of a DC current via the electrodes and the detection of the voltage drop produced between the electrodes due to this current flow. Since, however, the GSR is usually recorded for prolonged periods, the electrodes used cannot be relied upon as these electrodes will undoubtedly produce a DC offset potential of several hundred millivolts due to the passage of current for a prolonged period. This offset potential could not be differentiated from a potential produced by a change in the subject's resistance. For this reason, DC techniques have been found to be unsatisfactory.

AC
techniques

Since it is desirable to employ a measurement technique sensitive primarily to resistance change, and since it is undesirable to use DC measurement techniques, very low frequency AC techniques are invariably used. These measurement techniques involve the passage of an AC current of perhaps 10 microamperes peak at a frequency of 2 or 3 Hz. The resulting voltage drop between the electrodes can then be detected as an AC signal which will be independent of any DC offset potentials generated at the electrodes; such a technique is shown in Fig. 13-2. Referring to Fig. 13-2, a constant-current pulse generator adapter, used in conjunction with a Type 561B or 564B oscilloscope's calibrator, provides pulses of 10 microamperes, with a duration of 40 milliseconds. Further details on this adapter are given in Chapter 29. Since this pulse waveform has only a 10% duty cycle, a change in subject resistance, when using an AC-coupled amplifier, will primarily alter the displayed pulse amplitude and will cause almost no shift in the oscilloscope zero level or base line.

electrodes

Since, when recording the GSR, we are recording resistance changes due to action of the sweat glands at the surface of the skin, electrodes should be used that make direct contact with the skin as the use of any conductive electrode paste would interfere with the action of the sweat glands. The electrodes used should also have no chemical effect on the action of the sweat glands. Thin lead plates are preferred for GSR electrodes as they meet the above requirements and they are also malleable and can be molded to suit the subject's contour. Lead plate electrodes are used in the photograph shown in Fig. 13-2; one electrode is placed on the palm of the hand in a region of high sweat gland concentration and the other electrode is placed on the back of the hand in a region almost completely devoid of sweat glands.

typical resistances

The range of resistances encountered in normal subjects when recording between two electrodes on the hand is from 20,000 ohms to perhaps 0.2 megohms. In some instances, particularly if the autonomic nervous system is malfunctioning and the sweat glands are effectively denervated, the resistance between the electrodes will exceed 1 megohm.

13.3 GSR MEASUREMENT BY POTENTIAL DETECTION

offset potential

GSR measurement by this technique involves the detection of a DC potential between two electrodes on the subject. This DC potential will normally be less than one millivolt. An electrode offset potential in excess of one millivolt may be produced at the electrode/subject interface and any unbalance in this offset potential between the two electrodes cannot be differentiated from the GSR potential. Although solid silver electrodes, or perhaps silver plate electrodes, are used, this offset potential unbalance is difficult to control and will invariably contribute a considerable DC potential to the GSR potential. For this reason, GSR measurement by this technique is rarely attempted.

13.4 ELECTRICAL SKIN RESISTANCE

The electrical skin resistance (ESR) is basically the same as the GSR. The term ESR is, however, usually reserved for measurement of the distribution of sweat glands on the human body rather than the actual change in the activity of these sweat glands. The ESR is measured in the same way as the GSR, however the ground electrode is applied by means of a silver clip attached to the ear and the active or exploring electrode consists of a noncorroding silver or lead disc or a small roller wheel. Sweat gland distribution is detected by moving the position of this exploring electrode. The change in resistance between a sweating and a nonsweating area is distinct and a small movement of the exploring electrode can result in a resistance change in excess of 100 percent. Instruments specifically designed for measurement of the electrical skin resistance are referred to as dermometers.

sweat
gland
distribution

dermometer

ULTRASONOGRAPHY

acoustical
reflections

Ultrasonography is a technique by which ultrasonic energy is used to detect internal body organs. Bursts of ultrasonic energy are transmitted from a transducer through the skin and into the internal anatomy. When this energy strikes an interface between two tissues of different acoustical impedance, reflections are returned to the transducer. The transducer converts these reflections to an electrical signal. This electrical signal is amplified and displayed on an oscilloscope, each tissue interface appearing as a vertical deflection along the baseline of the oscilloscope at a distance proportional to the depth of the interface. This ultrasonic technique is similar to the time-domain reflectometry technique used to measure electrical cable length and the sonar technique used to detect objects under water.

advantages

While the use of pulse-echo ultrasonic energy is somewhat similar to the use of X-ray, the results obtained differ from an X-ray picture, being a cross-sectional projection or simply a linear projection rather than a profile of the area examined. Also, in contrast to X-ray, ultrasonography uses mechanical energy at a level which is not harmful to human tissue, thus, it may be used with safety on pregnant subjects and for frequent examination. Ultrasonography can detect materials that are not radiopaque, thus angiographic dyes are unnecessary. As commercial ultrasonic diagnostic instruments are easy to operate, ultrasonography is rapidly becoming a valuable diagnostic technique.

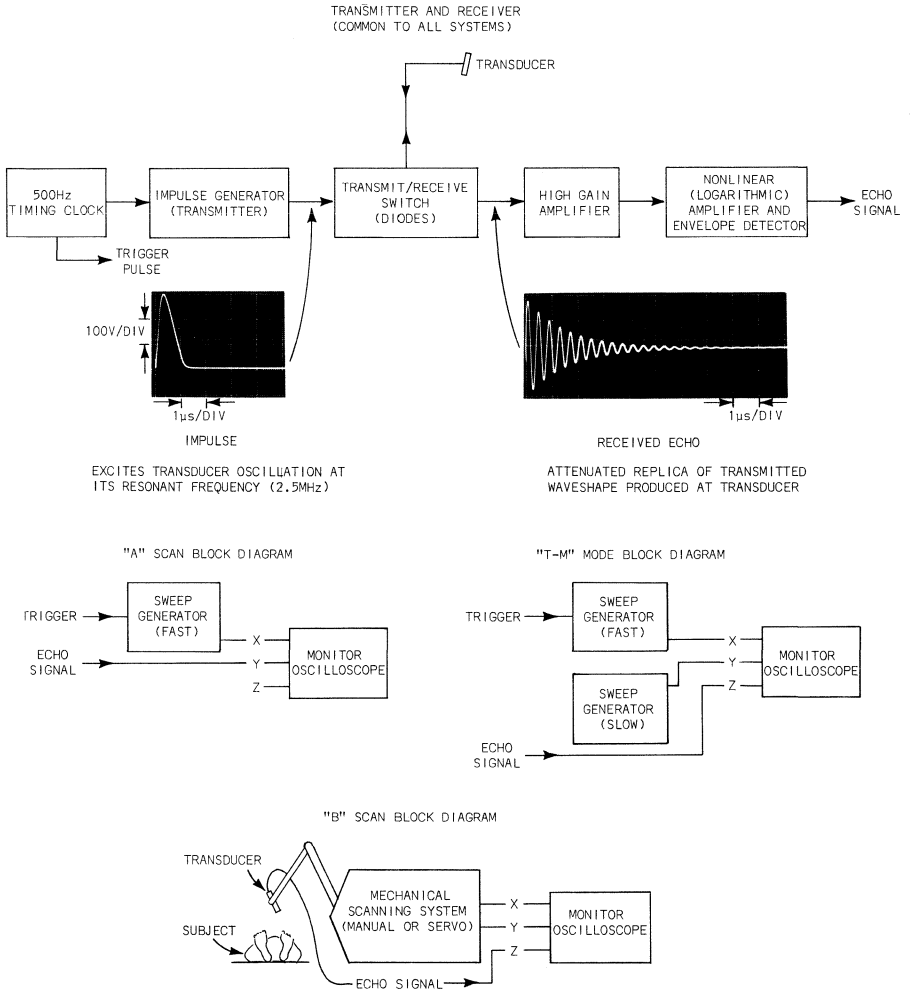


Fig. 14-1. Ultrasonic diagnostic systems -block diagrams.

display
modes

While the basic function of ultrasonic diagnostic equipment is to measure distances between interfaces that separate body structures by timing the echos produced by these interfaces, this timed-echo information may be processed in various ways to produce different forms of display. The three commonly used display modes are the "A" scan mode, the "T-M" mode and the "B" mode; further details on these three operating modes are given later in this chapter.

14.1 ULTRASONIC SYSTEMS

transducer
frequencies

Ultrasonic energy for use in ultrasonography is produced by exciting a piezoelectric crystal referred to as the ultrasonic transducer. These piezoelectric crystals normally have a self-resonant frequency of between 1 MHz and 10 MHz; the most common type used for ultrasonography having a self-resonant frequency of 2.5 MHz. When this crystal transducer is excited by an electrical impulse, it will ring at its self-resonant frequency and produce a train of damped oscillations. Fig. 14-1 shows the input pulse used to excite a 2.5 MHz transducer and a reflected pulse detected by the same transducer. The input pulse has an amplitude of almost 300 volts and a duration of 1 microsecond. The transducer oscillates at 2.5 MHz, the damping factor being dependent on the transducer design and on the type of tissue in the transducer's path. As this transmitted damped oscillation reaches an interface between materials having different acoustical impedances, a reflection or echo is produced. By the time this echo has returned to the transducer, the transducer is passive and is then again excited by vibrations produced by this returning echo. This echo signal is then amplified and processed in a logarithmic amplifier and an envelope detector. This process is repeated at an approximate 500 Hz rate. The time between the generation of the transducer exciting pulse and the detection of the received echo represents the time taken for the ultrasonic energy to travel from the transducer, to the interface, and thence from the interface back to the transducer. The velocity of sound waves, and thus of ultrasonic energy, in body tissue is about 0.125 centimeters per microsecond.

operational
theory

sound
velocity

It is rarely necessary to know this velocity accurately as relative interface distances are more important than actual interface distances. If these actual distances are required, the system can be calibrated against a known tissue/interface distance.

depth
versus
resolution

Depth and resolution of the system depends on the crystal's resonant frequency, a 1 MHz crystal provides low resolution, however, reflections can be detected for tissue 50 centimeters from the transducer. The commonly used 2.5 MHz crystal can be used to at least 20 centimeters; whereas a 10 MHz crystal provides excellent resolution, however it cannot be used above about 5 centimeters. Relating distance to velocity, a tissue interface 5 centimeters from the transducer will produce a reflection 80 microseconds after the impulse.

transducer/
tissue
coupling

It is important that the transducer be firmly held against the tissue (skin) as any air in the transducer's path will severely attenuate the ultrasonic energy. A satisfactory transducer/tissue coupling can be assured by using a liquid coupler such as water. In certain ultrasonic scanning systems it is impossible to locate the transducer against the tissue and a substantial depth of water is used as a coupling medium.

14.2 "A" SCAN ULTRASONOGRAPHY

Y-T mode

calibrated
sweep
generator

"A" scan ultrasonography displays the amplified echo signal on the vertical channel of an oscilloscope with the horizontal channel being deflected by a conventional sweep generator. This sweep generator is triggered from the impulse signal and the time delay between the beginning of the sweep and the echo appearing on the CRT screen is proportional to tissue depth. The sweep generator may be a conventional sweep generator or may be specifically calibrated in tissue depth (centimeters) rather than in time per division. A sweep speed of 100 microseconds per division would correspond to approximately 6.25 centimeters of tissue per division. Faster sweep speeds would give correspondingly greater resolution. The fastest usable sweep speed would probably be 2 microseconds per division which would correspond to 1.25 millimeters of tissue per division. This faster sweep speed would only be usable with 10 MHz transducers.

echoencephalography

The most common usage of "A" scan ultrasonography is in echoencephalography. Echoencephalography detects brain midline position and possible displacement of this midline due to an abnormal space occupying mass within one side of the skull, such as a tumor. When performing an "A" scan with the transducer held against the side of the subject's head, echos are received from the two halves of the brain as shown in Fig. 14-2. The echo produced by the midline is termed the "M" echo and should be symmetrically placed between the echos received from each side of the subject's skull.

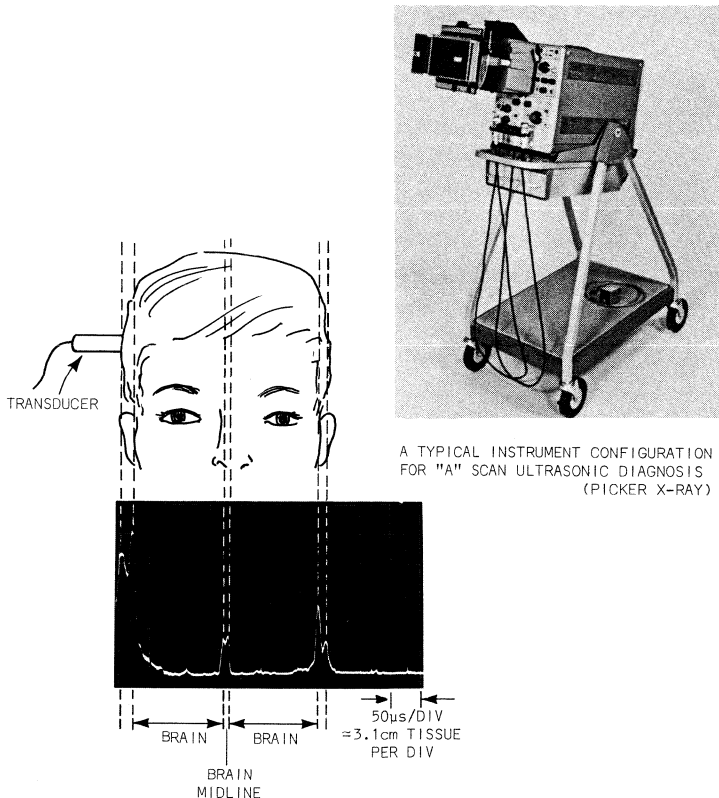


Fig. 14-2. "A" scan echoencephalography.

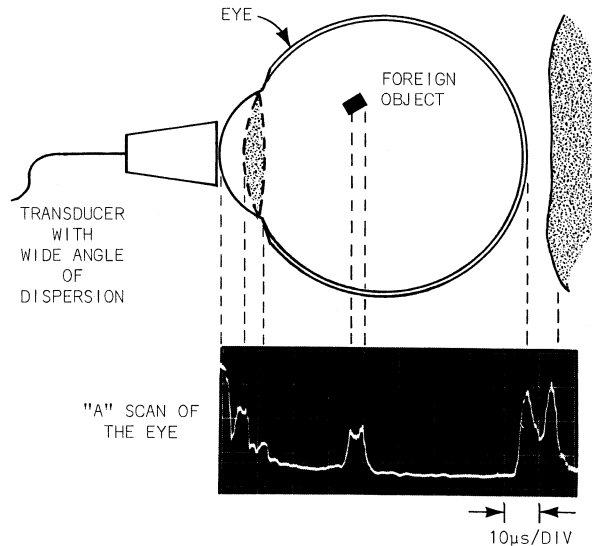


Fig. 14-3. "A" scan detection of a foreign object.

foreign
object
detection

"A" scan ultrasonography may also be used to detect foreign objects imbedded in a subject. Fig. 14-3 shows a typical "A" scan obtained when scanning an eye containing a foreign object with a 10 MHz transducer. The response attributed to the foreign object can be verified by scanning the other eye and noting the lack of any response at this time position.

Many manufacturers produce equipment suitable for "A" scan ultrasonography. A typical instrument configuration is shown in Fig. 14-2; a Tektronix Type 561B or 564B Oscilloscope is used with a Tektronix Type 2B67 time-base unit and a special vertical plug-in unit, designed for ultrasonography, produced by the Picker X-Ray Corporation. Suitable probes are also produced by the Picker X-Ray Corporation. The complete instrument may be used in conjunction with a Tektronix Scope-Mobile® Cart and Tektronix Trace-Recording Camera.

14.3 "TIME-MOTION" MODE ULTRASONOGRAPHY

Z-T mode

Considering "A" scan ultrasonography as discussed previously; if an interface producing an echo was moving in relation to the transducer, the response obtained on the CRT would also move horizontally relative to the beginning of the sweep. If this "A" scan system was to be modified slightly by removing the echo signal from the vertical channel (Y) and connecting it to the intensifying channel (Z), an echo response would then appear as an intensified spot on the screen. Any "motion" in this echo response would appear as a horizontal motion of this spot between successive sweeps of the horizontal sweep generator. No vertical signal would be involved.

In time-motion or "T-M" mode ultrasonography, the above hypothetically modified "A" scan system is used with a slow sweep generator moving the display vertically. Thus, any motion in the echo response is displayed in real time by the slow vertical sweep. This system is shown diagrammatically in Fig. 14-1, and a typical "T-M" mode ultrasonogram from motion of the mitral valve of the heart is shown in Fig. 14-4.

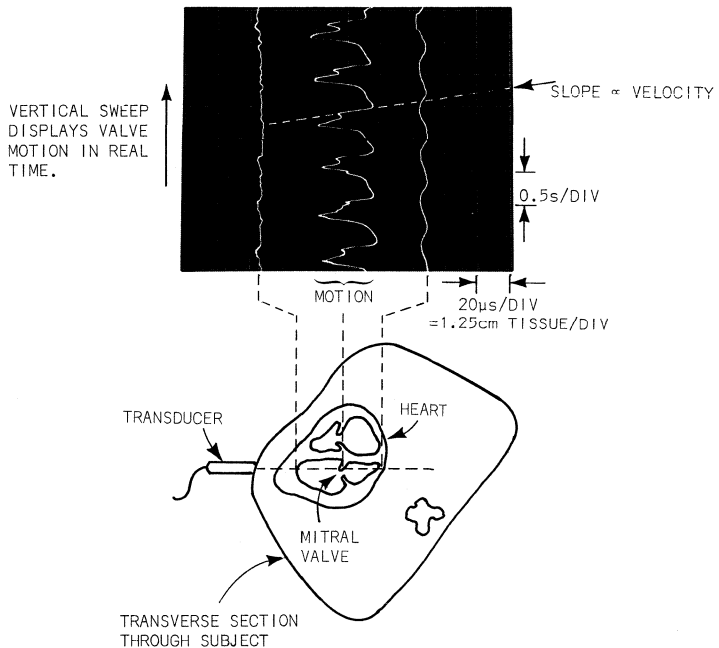
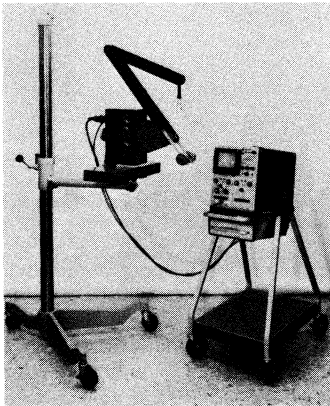
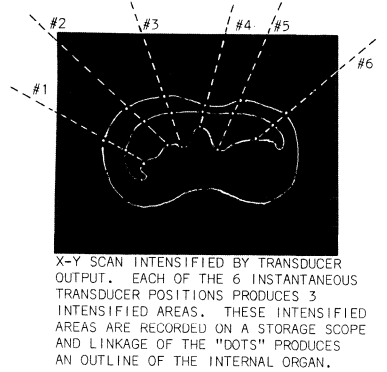
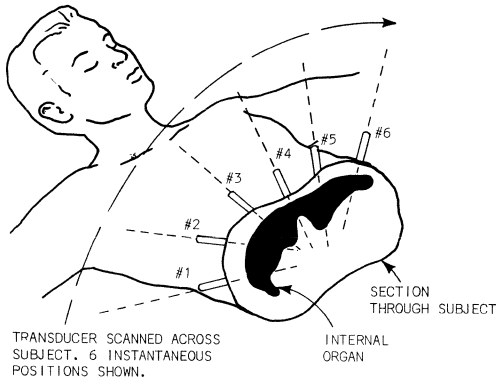
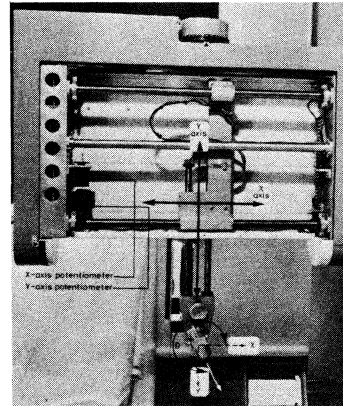


Fig. 14-4. "T-M" mode cardiography.



A TYPICAL MANUAL SCANNING SYSTEM PRODUCED BY THE PICKER X-PAY CORP., LONGMONT, COLO.



TYPICAL $XY\theta$ AUTOMATIC SCANNING SYSTEM. THE POSITION OF THE TRANSDUCER IS PRESENTED AS FOUR VOLTAGES, REPRESENTING THE DISTANCES ALONG THE X AND Y AXES AND THE SINE AND COSINE OF THE ANGLE BETWEEN THE LINE OF THE ULTRASONIC BEAM AND THE X-AXIS.

Fig. 14-5. Ultrasonic scanning - "B" mode.

cardiography

Referring to Fig. 14-4, the motion of the mitral valve is clearly shown on the vertically-swept trace on the face of the CRT. Fig. 14-4 also shows the motion of the walls of the heart, however echoes from other interfaces in the transducers path have been gated out in this instance. This particular application, known as "T-M" mode cardiography is particularly valuable in the detection of mitral stenosis. Since the actual movement of the mitral valve is shown in centimeters in real time (vertical sweep), the velocity of movement of this valve can be determined from the slope of the trace obtained. In the example shown in Fig. 14-4, the maximum velocity of the mitral valve is 16 centimeters per second, the overall movement of the valve is 2.2 centimeters and the heart rate is 86 beats per minute.

14.4 ULTRASONIC SCANNING

mechanical
X-Y scan

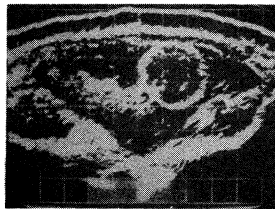
Ultrasonography using the "B" mode technique is referred to as ultrasonic scanning. In this mode, the echo signal is connected to the Z axis of a monitor to provide intensification as in the "T-M" mode, however the X and Y monitor input signals are derived from a mechanical scanning system and provide signals proportional to the position and direction of the probe to form part of a physical picture of the organs being examined. The result obtained is a two-dimensional, cross-sectional presentation of part of the subject.

manual
scanning

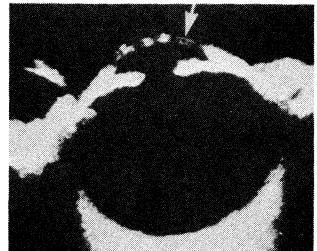
In its simplest form the "B" mode ultrasonic scanner is manually scanned by an operator across the part of the subject of interest. Referring to Fig. 14-5, with the transducer placed in each of the six positions shown, a series of intensified areas appears on the CRT face. Their position bearing an exact physical relationship to the positions of tissue interface within the subject. With only six transducer positions as shown in Fig. 14-5, only six rows of intensified dots would appear on the CRT, however, in practice, the transducer is slowly scanned over the subject and is producing echo patterns at a 500 Hz rate. The display thus appears continuous rather than composed of several dots. The manual scanning system shown in Fig. 14-5, while not offering the sophistication of an automatic scanning system, is economical and may also be used for both "A" mode scanning and "T-M" mode scanning rather than being restricted to "B" mode use.

automatic
X-Y- θ
scanning

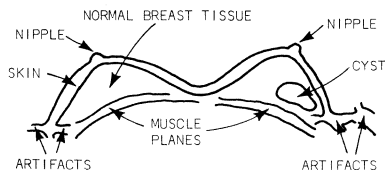
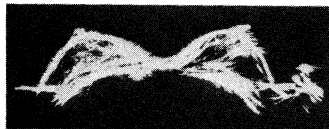
The mechanism associated with an automatic X-Y- θ compound scanning system is also shown in Fig. 14-5. Automatic compound scanning systems can be divided into two distinct classes, in one class the scanning movements of the probe are predetermined and related to the geometry of the machine itself, in the other the scanning pattern is dictated by the contours of the particular part of the individual patient being examined. The first type is referred to as an immersion scanner since a liquid-filled space is required between the probe and the subject and the second type is referred to as a dry scanner since the transducer is in direct contact with the subject. The mechanism associated with immersion scanners is somewhat less complicated than the mechanism required for direct contact dry scanners and the requirement of having a liquid filled space is satisfied by having a plastic membrane separating the subject from the liquid in which he or she is immersed. Immersion scanners are particularly suited to areas such as the breasts and the eyes where direct contact scanning is impractical.



POSITION OF FETUS *IN UTERO*



CORNEAL THICKNESS DETERMINATION



DETECTION OF CYST IN BREAST TISSUE

Fig. 14-6. Typical "B" mode displays from a compound scanning system.

ultra-
sonographic
applications

Fig. 14-6 shows typical "B" mode displays obtained with a compound scanning system. These displays are self-explanatory and indicate some of the potential advantages of ultrasonography. Neither of the three displays shown in Fig. 14-6 could have been obtained by conventional X-ray techniques as; for the fetus in utero, fetal development is impaired by X-ray energy; for the breast cyst, such cyst tissue is undetectable by conventional X-ray techniques; and, for the scan of the eye, resolution with X-ray techniques cannot be obtained if the X-ray plate cannot be located directly behind the region concerned.

14.5 DOPPLER ULTRASOUND

movement
causes
frequency
shift

Continuous ultrasonic energy, rather than bursts of ultrasonic energy as used in ultrasonography, is used to detect motion within a subject by the Doppler principle. When ultrasonic energy is reflected from a moving object it is shifted slightly in frequency, the frequency shift being proportional to the speed of the object. In the living body there are numerous movements which reflect ultrasonic energy: blood flowing through arteries, the action of the heart, intestinal movements and passage of urine and gastric juices. The most common application of Doppler ultrasonics is in obstetrics to detect movement of the fetal heart and fetal blood flow; such fetal activity can be detected as early as the tenth week of gestation. Another major application for Doppler ultrasonics is in the detection of blood flow in the peripheral circulation of the body.

SECTION III

INSTRUMENTATION

The following chapters ([5-]8) describe the characteristics of the instrumentation required to implement the measurement techniques covered in Section II as well as the characteristics of instrumentation in current use for biophysical measurements. A working knowledge of electronic instrumentation principles is assumed; this material primarily discusses the instrumentation characteristics that are unique to the biophysical sciences.

Since the following material is generally limited to a discussion of those instrumentation characteristics that are unique to biophysical measurement, it should not be regarded as a general reference source on any one type of instrumentation. An attempt is made in each of the following chapters to direct a reader who is somewhat unfamiliar with general electronic instrumentation to a general reference source. This general reference source should first be studied by such readers before proceeding with a study of the material presented in the following chapters.

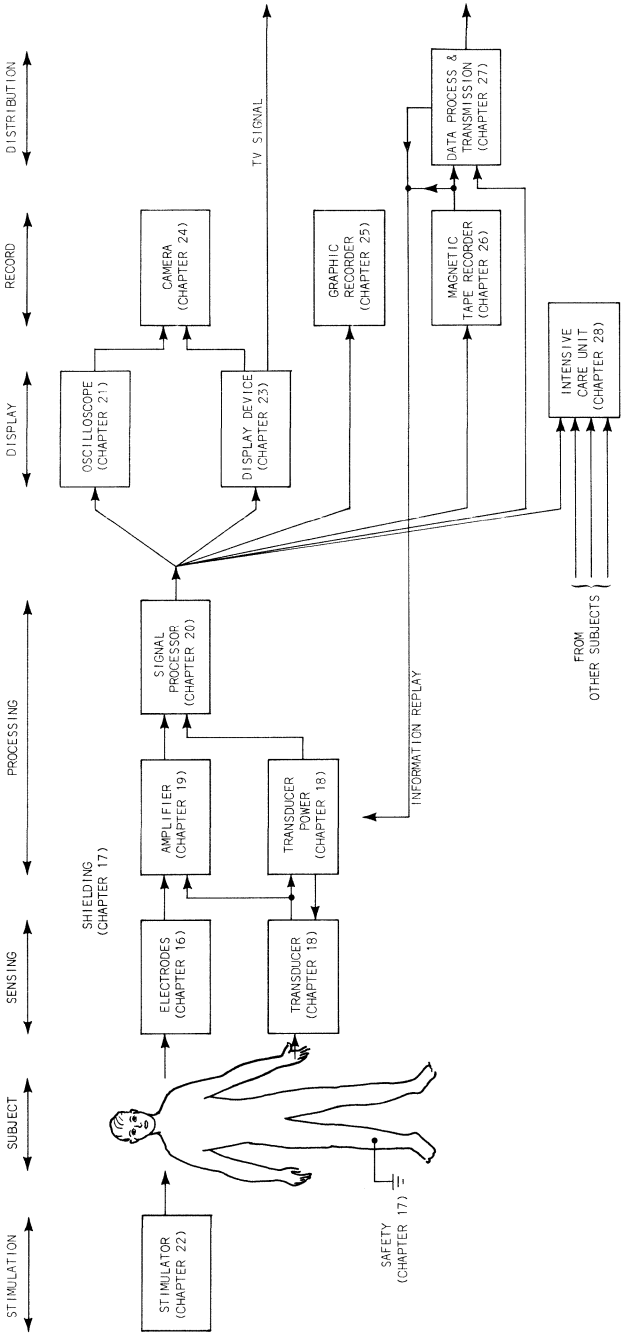


Fig. 15-1. Biomedical instrumentation configuration (Showing references to appropriate chapters in this book).

15

INSTRUMENTATION SUMMARY

A block diagram of the instrumentation used for biomedical measurements is shown in Fig. 15-1. While the majority of measurement systems do not utilize all of the components shown in this diagram, the ultimate measurement system may contain all components. The following 13 chapters (Chapters 16-28) describe each of these instrumentation groups in more detail.

biomedical
signal
source

The biomedical signal to be measured is derived from the subject. This signal may be continuously produced by the subject (such as the ECG), it may be voluntarily initiated by the subject (such as a muscular contraction) or it may be produced by artificial stimulation of the subject (such as an evoked response). This artificial stimulation is normally produced by an electric current; however, other forms of stimulators, such as mechanical or optical stimulators, are occasionally used. Stimulators are covered in more detail in Chapter 22.

electrodes
and
transducers

The biomedical signal is either directly available from the subject as an electrical potential detected by using electrodes or it is available in some other form, in which case an electrical potential corresponding to the biomedical signal is produced with the aid of a transducer. Biomedical measurements such as body impedance are regarded as transducer-type measurements (although electrodes are used to perform this measurement) as the measurement involves the detection of an impedance rather than the detection of a bioelectric potential. Electrode systems are covered in more detail in Chapter 16 and transducer systems are covered in more detail in Chapter 18. Electrode shielding and subject safety is covered in Chapter 17.

amplifying
and
processing

Whether the biomedical signal is obtained directly via electrodes or produced via transducers, it invariably requires amplification and perhaps signal processing. Amplifiers are discussed in Chapter 19 and signal processes are discussed in Chapter 20. Many transducer systems incorporate the necessary amplification into the system as discussed in Chapter 18.

displaying

Once the biomedical signal has been amplified and processed, then biomedical measurement techniques can be forgotten as electronic measurement and display techniques are now employed. The amplified and processed signal may be displayed on a conventional oscilloscope (Chapter 21), on some other form of display device such as a television receiver via a scan converter (Chapter 23), on a paper chart recorder (Chapter 25) or it may be recorded on magnetic tape (Chapter 26). The signal may be directly routed to further data processing and data transmission equipment (Chapter 27), or it may be combined with signals from other patients in an intensive care unit (Chapter 28).

permanent
record

If the signal is recorded on magnetic tape or on a chart recorder, then a permanent record of the signal is obtained. When monitoring the signal on a conventional oscilloscope or a display device, a permanent record must be obtained by photographic methods as covered in Chapter 24. Use of a magnetic tape recorder or data processing and transmission equipment allows information replay through the instrumentation system for a detailed analysis of biomedical signals and, in many cases, frees the operator to concentrate on recording the biomedical information.

15.1 BIOMEDICAL SIGNALS

Fig. 15-2 lists the biomedical measurements covered in Section II of the book. This summary is necessarily very brief and references to the appropriate sections in Section II are given to direct the reader to more detailed information. Fig. 15-3, produced by courtesy of Beckman Instruments, Inc., is a more complete summary of biophysical measurements and gives the ranges and characteristics of the various biomedical signals. (Note: Not all of the biomedical measurements covered in Fig. 15-3 are included in this book.)

PHYSIOLOGICAL PARAMETER	REFERENCE (THIS BOOK)	MEASUREMENT REQUIRED	SENSING DEVICES USED	PARAMETER TYPICAL P-P VALUE	PARAMETER FREQUENCY CHARACTERISTICS	
					FUNDAMENTAL Hz	SPECTRUM F(ω) Hz
<u>CIRCULATORY SYSTEM</u>						
HEART POTENTIALS	CHAPTERS 2 AND 5	ELECTROCARDIOGRAM	SURFACE ELECTRODES	2mV	1.3	.05 - 80
	CHAPTER 17		HEART ELECTRODES	50mV	1.3	.05 - 80
	CHAPTER 6	VECTORCARDIOGRAM	SURFACE ELECTRODES	2mV	1.3	.05 - 80
	CHAPTER 7	FETAL ELECTROCARDIOGRAM	SURFACE ELECTRODES (MOTHER)	10 μ V	2.5	2 - 100
BLOOD PRESSURE	SECT 8.1	DIRECT ARTERIAL PRESSURE AT BRACHIAL ARTERY OR FEMORAL ARTERY	PRESSURE TRANSDUCER	120mmHg	1.3	DC - 20
	SECT 8.1	DIRECT VENOUS PRESSURE	MERCURY MANOMETER	120mmHg	"	"
	SECT 8.1		PRESSURE TRANSDUCER	9mmHg	1.3	DC - 20
	SECT 8.2	INDIRECT ARTERIAL PRESSURE	WATER MANOMETER	120cmH ₂ O	1.3	-
	SECT 8.3	RELATIVE ARTERIAL PRESSURE	SPHYGMOMANOMETER WITH KOROTKOFF MICROPHONE	120/80mmHg	-	-
			PLETHYSMOGRAPH	150mV	1.3	30 - 500
	SECT 8.4	PERIPHERAL FLOW	IMPEDANCE PLETHYSMOGRAPH	RELATIVE	1.3	.05 - 10
BLOOD FLOW (AND BLOOD VELOCITY)			ELECTROMAGNETIC FLOWMETER	0.1% CHANGE	1.3	.05 - 10
			ISOTHERMAL FLOWMETER	1000cc/min	1.3	DC - 50
			ELECTROTURBINOMETER	"	"	"
			ULTRASONIC FLOWMETER	"	"	"
	SECT 8.5	CARDIAC OUTPUT	FLOWMETER AT AORTA	5000cc/min	1.3	DC - 50
			DYE DILUTION	5000cc/min	-	-
BLOOD VOLUME	SECT 8.6	BLOOD VOLUME	MODIFIED DYE DILUTION	5500cc	-	-
MITRAL VALVE FUNCTION	SECT 14.3	ULTRASONIC ECHO	T-M MODE ULTRASONICS	16cm/s	1.3	DC - 50
<u>RESPIRATORY SYSTEM</u>						
BREATHING	CHAPTER 9					
	SECT 9.2	PNEUMOGRAM	THERMISTOR PNEUMOGRAPH	500cc/BREATH	0.25	.05 - 2
			IMPEDANCE PNEUMOGRAPH	"	"	DC - 2
			ELASTIC FORCE GAGE	"	"	"
RESPIRATORY FLOW	SECT 9.3	PNEUMOTACHOGRAM	PNEUMOTACHOGRAPH WITH PRESSURE TRANSDUCER	20,000cc/min	"	"
RESPIRATORY VOLUME	SECT 9.4	SPIROGRAM	SPIROMETER	4000cc	"	DC - .5
<u>BRAIN FUNCTIONS</u>						
ELECTRICAL ACTIVITY	CHAPTER 10	ELECTROENCEPHALOGRAM	SCALP ELECTRODES	50 μ V	10	.5 - 100
			INTRACRANIAL ELECTRODES	500 μ V	"	"
EVOKED RESPONSES	CHAPTER 11	INTRACELLULAR POTENTIALS	MICROELECTRODES	100 μ V	"	1 - 10,000
		EXTRACELLULAR POTENTIALS	NEEDLE ELECTRODES	50 μ V	"	1 - 1000
EYE RESPONSES	SECT 11.8	ELECTRORETINOGRAM	CONTACT LENS ELECTRODE	100 μ V	"	.05 - 20
BRAIN MIDLINE POSITION	SECT 14.2	ULTRASONIC ECHO	"A" SCAN ULTRASONICS	"	"	"
<u>MUSCULAR FUNCTIONS</u>						
MUSCLE EXCITABILITY	CHAPTER 12					
	SECT 12.3	S-D CURVE	STIMULATE WITH SURFACE ELECTRODES	"	"	"
MUSCLE STRENGTH	SECT 12.4	MYOGRAM	NEEDLE OR SURFACE ELECTRODES	300 μ STRAIN	.5	DC - 50
MUSCLE POTENTIALS	SECT 12.6	ELECTROMYOGRAM	NEEDLE OR SURFACE ELECTRODES	1mV	"	10 - 5000
	SECT 12.7	ELECTROMYOGRAM WITH STIMULATION	NEEDLE OR SURFACE ELECTRODES STIMULATE WITH SURFACE ELECTRODES	"	"	"
NERVE CONDUCTION	SECT 12.8	H REFLEX RESPONSE	AS ELECTROMYOGRAPH WITH REDUCED STIMULATION	"	"	"
	SECT 12.9	CONDUCTION VELOCITY	AS ELECTROMYOGRAPH	"	"	"
SMOOTH MUSCLE ACTIVITY	SECT 12.11	ELECTROGASTROGRAM	SURFACE ELECTRODES	20mV	.25	.05 - 2
<u>AUTONOMIC NERVOUS SYSTEM</u>						
SWEAT GLAND ACTIVITY	CHAPTER 13					
		GALVANIC SKIN REFLEX	LEAD SURFACE ELECTRODES	50k Ω	"	"
		ELECTRICAL SKIN RESISTANCE	" " "	"	"	"
BODY TEMPERATURE	SECT 9.5	TEMPERATURE	THERMISTOR PROBE THERMOMETER	98°F	"	"
<u>ANATOMY</u>						
INTERNAL ORGAN POSITION	CHAPTER 14	ULTRASONIC ECHO	ULTRASONIC SCANNING SYSTEM	"	"	"

Fig. 15-2. Biophysical measurements covered in this book.

PRIMARY SIGNAL RANGES AND CHARACTERISTICS	
<p><u>CARDIOVASCULAR SYSTEM</u></p> <p>BLOOD PRESSURE, DIRECT METHOD</p> <p>BLOOD PRESSURE, INDIRECT METHOD, INTERMITTENT SYSTOLIC AND DIASTOLIC</p> <p>PULSE WAVES, DIRECT METHOD, ARTERIAL</p> <p>PULSE WAVES, INDIRECT METHOD, PERIPHERAL ARTERY</p> <p>PHONOCARDIOGRAM</p> <p>PLETHYSMOGRAM (VOLUME MEASUREMENTS)</p> <p>BALLISTOCARDIOGRAM</p> <p>HEART RATE</p> <p>OXIMETRY</p> <p>CARDIAC OUTPUT</p> <p>BLOOD FLOW</p> <p>HEART SHUNT DETECTION</p> <p>ELECTROCARDIOGRAM</p>	<p>FREQUENCY RANGE: DC TO 200Hz; DC TO 60Hz USUALLY ADEQUATE. PRESSURE RANGE, ARTERIAL: 40 TO 300mmHg; VENOUS 0 TO 15mmHg.</p> <p>AUSCULTATORY CRITERION (KOROTKOFF SOUNDS): 30 TO 150Hz USUALLY ADEQUATE. PALPATORY CRITERION: 0.1 TO 60Hz. BOTH REQUIRE ADDITIONAL SIGNAL SHOWING OCCLUDING PRESSURE. (SEE BLOOD PRESSURE, DIRECT)</p> <p>FREQUENCY RANGE: 0.1 TO 60Hz USUALLY ADEQUATE. PULSE TRACE SIMILAR TO BLOOD PRESSURE, DIRECT, BUT WITHOUT BASELINE ZERO.</p> <p>FREQUENCY RANGE: 5 TO 2000Hz; MAJOR DIAGNOSTIC COMPONENTS LIE IN 20 TO 200Hz RANGE.</p> <p>FREQUENCY RANGE: DC TO 30Hz.</p> <p>FREQUENCY RANGE: DC TO 40Hz.</p> <p>AVERAGE RATE, HUMAN: 45 TO 200 BEATS/min; LAB ANIMAL; 50 TO 600 BEATS/min.</p> <p>FREQUENCY RANGE: 0 TO 60Hz; 0 TO 5Hz USUALLY ADEQUATE.</p> <p>FREQUENCY RANGE: 0 TO 60Hz; 0 TO 5Hz USUALLY ADEQUATE.</p> <p>FLOW RANGE, HUMAN: 5cc/min TO 10 LITERS/min. FREQUENCY RANGE: 0 TO 60Hz; 0 TO 20 USUALLY ADEQUATE.</p> <p>ELECTRODE SIGNAL RANGE, P_{aO_2}: 0 TO 160mmHg. HYDROGEN AND SODIUM ASCORBATE: QUALITATIVE.</p> <p>FREQUENCY RANGE: 0.05 TO 80Hz. SIGNAL RANGE: 10μV TO 5mV INCLUDES FETAL RANGE.</p>
<p><u>RESPIRATORY SYSTEM</u></p> <p>FLOW RATE (PNEUMOTACHOGRAM)</p> <p>BREATHING RATE CALCULATED FROM RECORD (WITH APPROXIMATE RELATIVE RESPIRATORY VOLUME)</p> <p>TIDAL VOLUME (MEASURED PER BREATH OR INTEGRATED TO PROVIDE VOLUME/min)</p> <p>CO₂, N₂O, OR HALOTHANE CONCENTRATION IN RESPIRED AIR</p> <p>DIFFUSION OF INSPIRED GAS (USING NITROGEN)</p> <p>PULMONARY DIFFUSING CAPACITY (USING CARBON MONOXIDE)</p>	<p>FREQUENCY COMPONENTS TO 40Hz. NORMAL FLOW RANGE: 250 TO 500ml/s; MAXIMUM 8 LITERS/s.</p> <p>AVERAGE RATE, HUMAN, 12 TO 20 BREATHS/min; LAB ANIMAL, 8 TO 60 BREATHS/min.</p> <p>TYPICAL VOLUME, ADULT HUMAN: 600ml/BREATH; 6 TO 8 LITERS/min.</p> <p>NORMAL RANGE, CO₂: 0 TO 10%; END-TIDAL CO₂, HUMAN: 4 TO 6%. N₂O: 0 TO 100%. HALOTHANE: 0 TO 3%.</p> <p>NORMAL RANGE OF NITROGEN CONCENTRATION DIFFERENTIAL: 0 TO 10%.</p> <p>NORMAL RANGE, HUMAN: 16 TO 35mlCO/mmHg/min.</p>
<p><u>DISSOLVED GASES AND pH</u></p> <p>PARTIAL PRESSURE OF DISSOLVED O₂, IN VIVO OR IN VITRO</p> <p>pH, IN VITRO</p> <p>PARTIAL PRESSURE OF DISSOLVED CO₂, IN VITRO</p>	<p>FREQUENCY RANGE: DC TO 1Hz USUALLY ADEQUATE. NORMAL MEASUREMENT RANGE: 0 TO 800mmHg P_{O₂}. HYPERBARIC P_{O₂} RANGE: 800 TO 3000.</p> <p>SIGNAL RANGE: 0 TO \pm700mV COVERS pH RANGE.</p> <p>NORMAL SIGNAL RANGE: 0 TO \pm150mV COVERS RANGE FROM 1 TO 1000mmHg P_{CO₂}.</p>
<p><u>BIOELECTRIC POTENTIALS</u></p> <p>ELECTROENCEPHALOGRAM</p> <p>CEREBRAL POTENTIALS, INTRACRANIALLY RECORDED</p> <p>ELECTROMYOGRAM (PRIMARY SIGNAL)</p> <p>ELECTROMYOGRAM (AVERAGED)</p> <p>SMOOTH MUSCLE POTENTIAL (e.g., ELECTROGASTROGRAM)</p> <p>ELECTRORETINOGRAM</p> <p>ELECTROCARDIOGRAM</p> <p>ELECTROWYSTAGMOGRAM</p>	<p>FREQUENCY RANGE: DC TO 100Hz; MAJOR DIAGNOSTIC COMPONENTS LIE IN 0.5 TO 60Hz RANGE. NORMAL SIGNAL RANGE: 15 TO 100μV</p> <p>NORMAL SIGNAL RANGE: 10μV TO 100mV. PULSE DURATION: 0.6ms TO 20ms.</p> <p>FREQUENCY RANGE: 10 TO 2000Hz. PULSE DURATION: 0.6ms TO 20ms.</p> <p>AN AVERAGE OF THE PRIMARY SIGNAL, AFTER FULL WAVE RECTIFICATION.</p> <p>FREQUENCY RANGE: DC TO 0.6Hz. NORMAL SIGNAL RANGE: 0.5 TO 80mV.</p> <p>FREQUENCY RANGE: DC TO 20Hz ADEQUATE. NORMAL SIGNAL STRENGTH: \pm1μV TO 1mV. (SEE LISTING UNDER CARDIOVASCULAR SYSTEM)</p> <p>DIRECT: FREQUENCY RANGE, 0 TO 20Hz. TYPICAL SIGNAL STRENGTH, 100μV/10° EYE MOVEMENT. DERIVATIVE OR VELOCITY: FREQUENCY RANGE, 0 TO 20Hz. SIGNAL DERIVED FROM DIRECT READINGS.</p>
<p><u>PHYSICAL QUANTITIES</u></p> <p>TEMPERATURE</p> <p>VOICE</p> <p>SKIN RESISTANCE (GSR)</p> <p>ISOMETRIC FORCE, DIMENSIONAL CHANGE, BODY FLUID AND BODY CAVITY PRESSURES</p>	<p>FULL RANGE OF SIGNALS.</p> <p>FREQUENCY RANGE: 20 TO 20,000Hz.</p> <p>RESISTANCE RANGE: 1k TO 500k.</p> <p>FULL RANGE OF SIGNALS.</p>

Fig. 15-3. Biophysical signal ranges

16

ELECTRODES

tissue-
electrode
interface

ionic-
electronic
currents

The electronics engineer, accustomed to measuring potentials between various points on an electronic circuit, may tend to regard two physiological electrodes simply as two probes applied to a subject in order to measure a potential difference. Such an oversimplification cannot be made as the conduction of current in tissue, as in any other liquid system, is ionic; that is to say, by the migration of positive and negative ions from point to point. To measure electrical effects in tissue it is necessary to make a transfer from this ionic conduction to the electronic conduction which occurs in the measuring circuit. This transfer is accomplished at the tissue-electrode interface.

16.1 ELECTRODE OFFSET POTENTIAL

half cell
potentials

If two dissimilar metals are inserted into a container of electrolyte, a potential difference can be measured between these metals. If one of these metals were silver and the other copper, this potential would be referred to as the silver/copper cell potential and would be in the order of 0.4 volts. This potential is comprised of two separate components added algebraically together: a "half cell" potential due to the silver electrode and a "half cell" potential due to the copper electrode. These half cell potentials, although difficult to measure, are the potentials produced across the metal-electrolyte interfaces. To a first order approximation, neglecting many chemical factors that would be important if the electrodes were used in a chemical measuring application rather than a biophysical measuring application, the half cell potential is approximately equal to the electrode potential of the metal concerned. Thus, the

electrode
potentials

METAL	IONIC SYMBOL	ELECTRODE POTENTIAL (EP)*
ALUMINIUM	Al ⁺⁺⁺	-1.66VOLTS
ZINC	Zn ⁺⁺	- .76
IRON	Fe ⁺⁺	- .44
LEAD	Pb ⁺⁺	- .12
HYDROGEN	H ⁺	0
COPPER	Cu ⁺⁺	+ .34
SILVER	Ag ⁺	+ .80
PLATINUM	Pt ⁺	+ .86
GOLD	Au ⁺	+1.50

CORRODED END
(ANODIC OR
LEAST NOBLE)

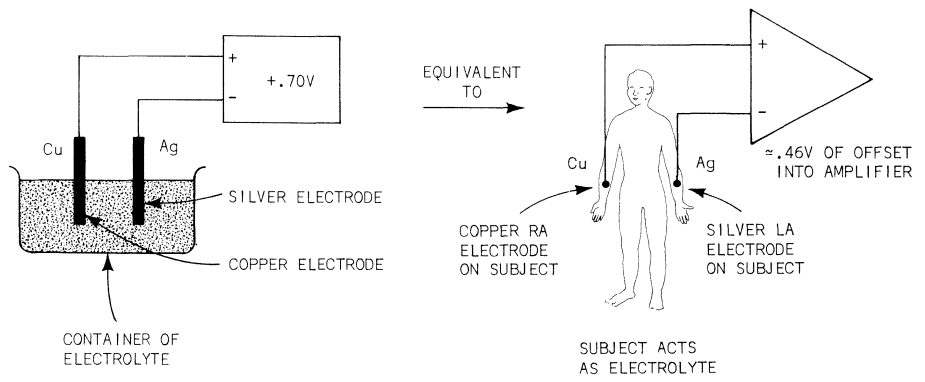
↑

↓

PROTECTED END
(CATHODIC OR
MOST NOBLE)

*APPROXIMATELY EQUAL TO HALF-CELL POTENTIAL
IN PHYSIOLOGICAL ENVIRONMENTS.

Fig. 16-1. Electrode potentials of metals, the electrochemical series.



OFFSET POTENTIAL PRODUCED BY DISSIMILAR METALS
APPROXIMATELY EQUALS DIFFERENCE IN ELECTRODE POTENTIALS.

$$\begin{aligned}
 \text{OFFSET} &\approx \text{EP Ag} - \text{EP Cu} \\
 &\approx 0.80\text{V} - 0.34\text{V} \\
 &\approx 0.46\text{V}
 \end{aligned}$$

Fig. 16-2. Electrode offset potential.

potential produced between the silver electrode and the copper electrode will be approximately equal to the difference in the electrode potentials of silver and copper, that is, approximately equal to 0.80 minus 0.34 or 0.46 volts. The exact value of this potential will depend on many chemical factors, the more important of which are the electrolyte used and the concentration of this electrolyte; however the value given is probably within about 100 millivolts of the actual value that may be measured in a physiological environment. A list of the electrode potentials for the more common metals is shown in Fig. 16-1. This listing is referred to as the electrochemical series. The example given of a silver and copper electrode is illustrated in Fig. 16-2, which shows the similarity between electrodes in a container of electrolyte and electrodes on a subject. In electrophysiology the difference in half cell potentials that can be detected between two electrodes is referred to as "offset potential."

electrode
offset
potential

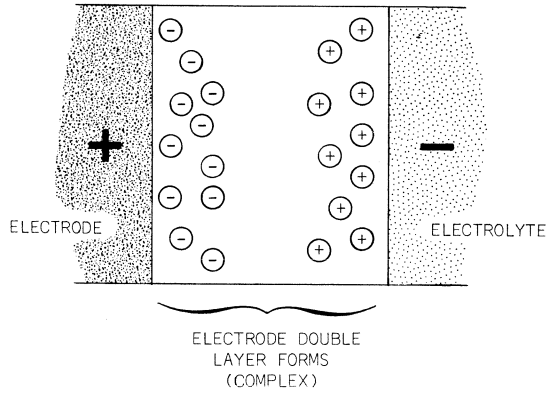
It is important not to confuse the electrode offset potential produced between two electrodes connected to a subject and the DC-differential offset potential available in many differential amplifiers. The DC-differential-offset-potential capability in differential amplifiers is intended to be used to cancel the electrode offset potential produced between electrodes.

differential
amp offset
potential

In the biomedical situation, the electrode offset potential produced between electrodes may be unstable and unpredictable. Thus, it is desirable that this potential be as low as possible. In order to reduce electrode offset potential it is first necessary to understand its origin. At any electrode/electrolyte interface there is a tendency for the electrode to discharge ions into solution and for ions in the electrolyte to combine with the electrode. These chemical reactions may be represented as follows:

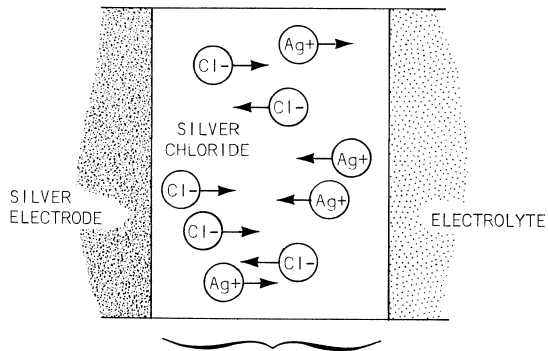
Metal \rightarrow electrons + metal ions
(Oxidization reaction)

Electrons + metal ions \rightarrow metal
(Reduction reaction)



POTENTIAL EXISTS BETWEEN THE ELECTRODE AND
ELECTROLYTE DUE TO THE FORMATION OF THE
ELECTRODE DOUBLE LAYER.

(A) METALLIC ELECTRODE



SILVER CHLORIDE FORMS FREE SILVER IONS (Ag^+)
AND CHLORIDE IONS (Cl^-) WHICH PREVENT THE
FORMATION OF THE ELECTRODE DOUBLE LAYER.

(B) SILVER/SILVER CHLORIDE ELECTRODE

Fig. 16-3. Electrode/electrolyte interface.

nonpolar-
izable
versus
polarizable

reversible
versus non-
reversible

partially
reversible
electrodes

The net result of these reactions is the creation of a charge gradient, the spatial arrangement of which is called the electrode double layer. This double layer may be represented, as shown in Fig. 16-3A, in its simplest form as two parallel areas of charge of opposite signs. Electrodes in which no net transfer of charge occurs across the metal-electrolyte interface are referred to as perfectly polarized or perfectly nonreversible electrodes, that is, only one of the two chemical reactions shown above can occur. Electrodes in which unhindered transfer of charge is possible are referred to as perfectly nonpolarizable or perfectly reversible electrodes, that is, both of the equations referred to above occur with equal ease. Practical electrode systems have properties that lie between these ideal limits. The electrode double layer existing between a practical electrode and electrolyte may be regarded as a battery in parallel with a reasonably high impedance as shown in Fig. 16-4. This impedance may typically be 10,000 ohms and the battery may be equivalent to a $10\ \mu\text{F}$ capacitor. The potential of the battery, or the charge on the capacitor, will be the half cell potential of the electrode.

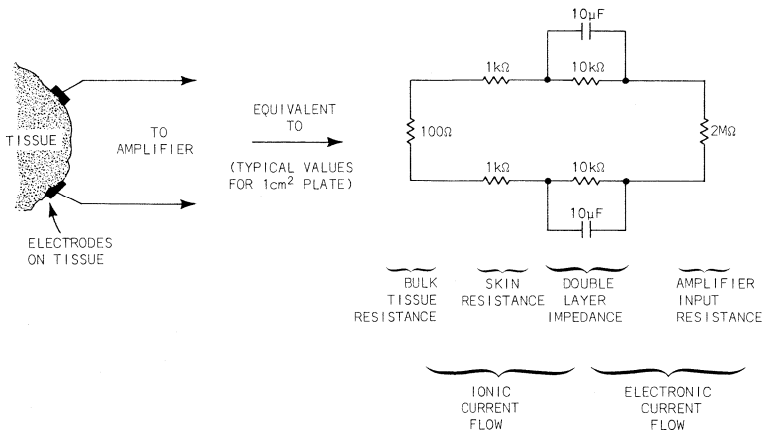


Fig. 16-4. Electrode equivalent circuit for partially reversible electrodes.

silver/
silver
chloride
electrodes

The silver/silver-chloride electrode shown in Fig. 16-3B consists of either a solid silver surface coated with a thin layer of solid silver chloride or, as is the case with Tektronix silver/silver-chloride electrodes, consist of silver powder and silver chloride powder compressed into a solid pellet. The presence of the silver chloride allows the electrode to behave as a near perfect nonpolarizable or reversible electrode as it prohibits the formation of the electrode double layer, the silver chloride dissociating to silver ions and chloride ions which are free to migrate between the electrode and the electrolyte, thus opposing the formation of the double layer. The net result is a low impedance, low offset potential, interface between the silver and the electrolyte. Both silver/silver-chloride and zinc/zinc-sulfate electrodes exhibited this characteristic.

toxicity

Most electrodes form soluble metallic salts and thus are highly toxic and can be used only as surface electrodes on intact skin. Silver electrodes are, however, nontoxic, as silver chloride is almost insoluble in a chloride-containing solution, thus very few free silver ions exist and tissue damage from them is negligible. Thus, silver or silver/silver-chloride electrodes are definitely preferred for use on exposed tissue. Although, as stated previously, zinc/zinc-sulfate electrodes produce low offset potential characteristics similar to silver/silver-chloride electrodes, they are highly toxic to exposed tissue due to the passage into the tissue of free zinc and/or sulfide ions.

chloriding

A layer of silver chloride may be deposited on silver electrodes, converting them to silver/silver-chloride electrodes, by a process known as chloriding. This is achieved by making the silver electrodes positive to a solution containing sodium chloride or saline (0.9% concentration) and passing a current through the electrode at the rate of 1 mA/cm^2 of surface for several minutes. The silver electrode should be cleaned to remove surface contaminants before chloriding.

jelly
(paste)
contact

Mechanical disturbance of the electrode double layer causes electrode noise because the double layer acts as a region of charge gradient, disturbance of which gives rise to a change in capacitance and thus a change in voltage. The electrical stability of an electrode is considerably enhanced by mechanical stabilization of the electrode-electrolyte interface. This is achieved by the use of indirect-contact floating electrodes which interposes an electrolyte "jelly" or "paste" between the electrode and the tissue.

The preceding discussion of electrode offset potential has been highly simplified, as in electrophysiology it is unnecessary to know the exact value of the electrode offset potential produced and it is therefore unnecessary for users of electrodes to understand all of the chemistry involved. Interested readers wishing to more fully understand the chemical process involved should refer to Geddes and Baker, *Principles of Applied Biomedical Instrumentation*, Wiley, 1968, Sections 9.2, 9.3, 11.1, 11.2 and 11.3 and also Dewhurst, *Physical Instrumentation in Medicine and Biology*, Pergamon, 1966, Section 21.2 (specifically Chapters 21, 22, 26 and 28).

16.2 ELECTRODE OFFSET POTENTIAL CHARACTERISTICS

drift and
noise

The previous section discusses the electrode offset potential produced at an electrode-electrolyte interface and states that this potential is unstable and unpredictable. When electrodes are connected to a subject in order to record a bioelectric event on a DC-coupled oscilloscope, long-term changes in electrode offset potential appear as baseline drift and short-term changes appear as noise on the trace.

discharging
offset

If a pair of new electrodes are applied to a subject, an electrode offset potential will exist as the electrodes can essentially be regarded as two batteries. When these electrodes are then connected to a DC-coupled oscilloscope amplifier, the input impedance of this amplifier will act as a load on the batteries and tend to discharge them. Since these electrodes form a very poor battery, a typical Tektronix oscilloscope differential-input impedance of two megohms will discharge the batteries over a period of several hours. It is recommended that new electrode pairs be shorted together in 0.9% saline for a few hours before use in an effort to accelerate this process. As these batteries are discharged their effective potential is reduced, this decreasing potential appearing as electrode offset potential drift as shown in Fig. 16-5. Although the source is represented as a battery in preceding discussions, it may equally well be represented as a capacitor with the amplifier differential-input impedance tending to discharge the capacitor as depicted in the equivalent circuit shown in Fig. 16-4. Either analogy is only a first order of approximation to the true situation.

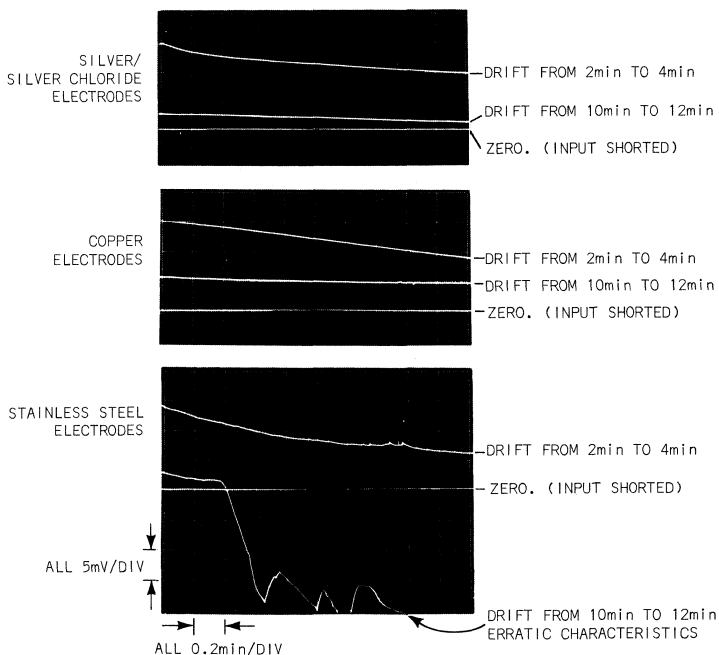


Fig. 16-5. Electrode offset potential long term stability-drift.

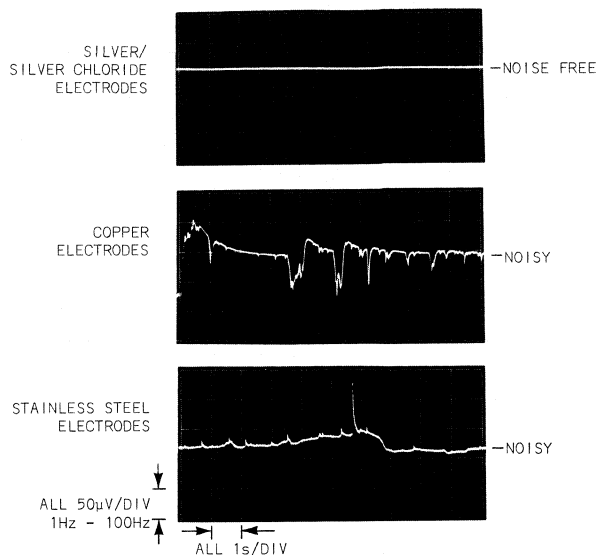


Fig. 16-6. Electrode offset potential short term stability-noise.

drift
comparison

Referring to Fig. 16-5, the electrode offset potential drift for silver/silver-chloride electrodes, copper electrodes and stainless steel hypodermic needle electrodes are shown. In each case the electrode offset potential drift is shown over a two-minute segment after the amplifier load had been applied to the electrodes for two minutes and for ten minutes. A "zero" trace is also included on each photograph. The electrode offset potential produced by the silver/silver-chloride electrodes and the copper electrodes appears to decrease exponentially; however, the stainless steel electrodes are clearly erratic and, for the particular pair of electrodes used for this investigation, show an abrupt change in potential after about ten minutes. The electrode offset potential drift for the silver/silver-chloride electrodes and the copper electrodes would be considered acceptable, however the erratic performance of the stainless steel electrodes may make them undesirable for physiological recording.

noise
comparison

Short term changes in electrode offset potential are referred to as electrode noise. If electrodes are used in conjunction with an AC-coupled high-sensitivity amplifier as shown in Fig. 16-6, the long-term drift characteristics will be rejected by the low frequency characteristic of the amplifier.

However, short-term instabilities are clearly displayed. From Fig. 16-6 it is clear that the silver/silver-chloride electrodes are essentially noise free, whereas copper electrodes and stainless steel electrodes exhibit noisy characteristics that may be intolerable for high-sensitivity physiological recording.

stored
charge

If current is passed through electrodes, this current will tend to charge the "battery" or "capacitor" formed at the electrode-electrolyte junction. If electrodes are used for tissue stimulation, this current flow is deliberate. However, any electrical activity associated with a subject, such as stimulation or defibrillation, will cause some degree of current flow in the recording electrodes. The ability of the electrodes to be charged like a battery or a capacitor by current flow is termed the electrodes' "offset-potential stored-charge characteristic." It is desirable that physiological electrodes store as low a charge as possible, thereby allowing an amplifier's differential-input impedance to discharge this energy in as short a time as possible. Referring to Fig. 16-7, silver/silver-chloride electrodes, copper electrodes and stainless steel electrodes were applied to a subject and used in conjunction with an amplifier having a 2-megohm differential-input impedance. A charge of 0.003 coulombs was then applied between these electrodes by discharging a capacitor through them. It can be seen from Fig. 16-7 that the silver/silver-chloride electrodes recovered to within 50 mV in 0.2 seconds and had almost completely recovered within 5 seconds. The copper electrodes exhibited inferior stored charge characteristics, taking 3.5 seconds to recover to 50 mV and several minutes to completely discharge. The stored-charge characteristics of these copper electrodes may be acceptable if no deliberate current were applied through the subject.

stored-
charge
comparison

The stainless steel electrodes exhibit particularly poor stored-charge characteristics, taking 120 seconds to recover to within 50 mV and in excess of six minutes to be sufficiently stable for ECG use. The stainless steel electrodes may cause erratic recording even if no current is deliberately passed through the subject since electrostatically and electromagnetically induced currents are always present.

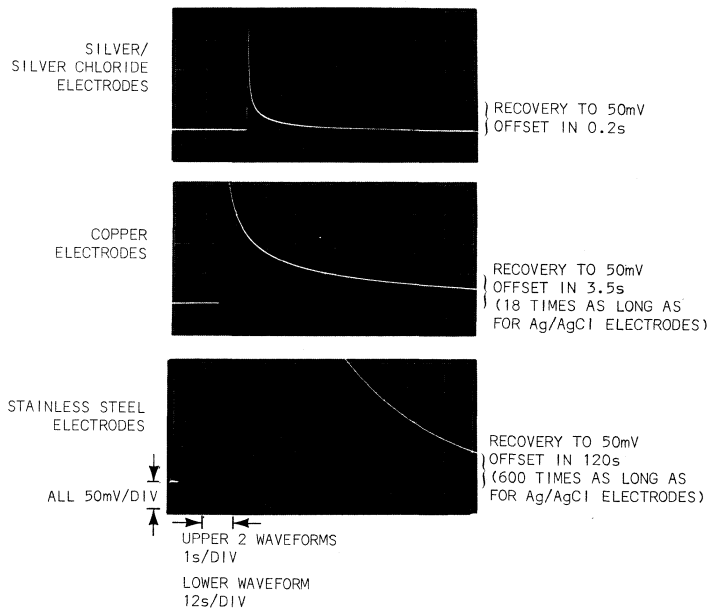
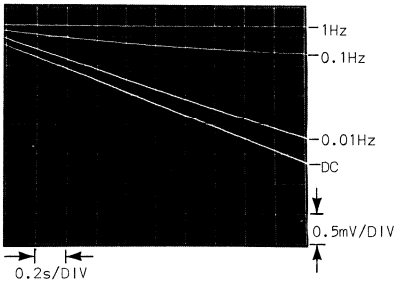


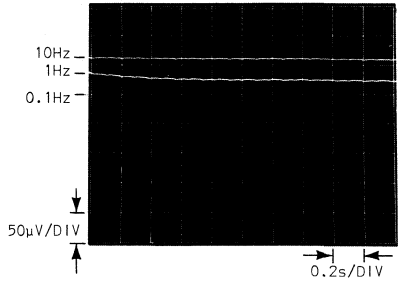
Fig. 16-7. Electrode offset potential stored charge after .003 coulomb surge - recovery.

EFFECT ON MEDIUM SENSITIVITY MEASUREMENT AT VARIOUS LOW FREQUENCY CUTOFF FREQUENCIES.



ELECTRODE DC OFFSET
DRIFTING AT 1mV/s.

EFFECT ON HIGH SENSITIVITY MEASUREMENT AT VARIOUS LOW FREQUENCY CUTOFF FREQUENCIES.



ELECTRODE DC OFFSET
DRIFTING AT 1mV/s.

Fig. 16-8. Electrode offset drift effect on AC coupled amplifier.

- drift and AC-coupling Long term electrode offset potential changes (drift) cannot simply be ignored when using AC-coupled instrumentation as the rate of change of this potential must be related to the AC-coupling time constant in the recording amplifier. Fig. 16-8 shows the baseline changes exhibited when using electrodes whose offset potential is drifting at a rate of 1 mV/s. At a sensitivity of 0.5 mV/div, as would be used to record the ECG, a 0.01-Hz low-frequency response offers little improvement over a DC-coupled instrument. When recording at high sensitivities, say 50 μ V/div as is the case with the EEG, this 1-mV/s electrode offset potential drift, used in conjunction with an amplifier having a 0.1-Hz low-frequency response, produces intolerable deflection and a 1-Hz low-frequency response must be used for satisfactory results. The drift rate of 1 mV/s chosen is typical of some of the poorer quality electrodes; however, silver/silver-chloride electrodes and many commercial electrodes can be expected to exhibit drift rates of between 10 μ V/s and 100 μ V/s after having been applied for more than two minutes.
- electrode offset versus amplifier maximum input specifications The absolute value of the electrode offset potential is rarely significant unless this absolute value exceeds the maximum differential-DC-input voltage characteristics of the amplifier concerned. When using DC-coupled instrumentation this electrode offset potential may exceed the maximum DC-differential offset available within the instrumentation and it would therefore be impossible to bring the trace onto the CRT screen.
- check 'unknown' electrodes If the characteristics of an electrode are unknown, it is recommended that tests be performed on the electrode before the electrode is used for physiological recording. Electrodes may be tested *in situ* or, alternatively, the subject may be replaced by a 0.9% concentration saline solution as were most of the electrode characteristics presented in this chapter.

16.3 OTHER ELECTRODE CHARACTERISTICS

- inertness The most important electrode characteristics are the characteristics associated with electrode offset potential, that is, drift, noise and recovery. Other factors must also be considered when selecting electrodes. The electrode must, of course, be a good conductor and must be chemically inert. Inertness, or the relative inactivity of the material in chemical reaction, is approximately proportional to the metal's electrode potential as shown in Fig. 16-1. Metals with a high negative electrode potential such as aluminum tend to be more chemically active than metals with a high positive electrode potential such as silver. If an electrode is not relatively inert, the formation of ions by the electrode material and the reactions between these ions and the acidity of the subject's skin produce poisons which can irritate the subject.
- mechanical character-
istics Mechanical characteristics of electrodes are important in many applications. Although it is of paramount importance that the electrode be satisfactory electrically, it must also be mechanically rugged, easy to clean and/or sterilize and easy to apply to the subject.
- photosensi-
tivity Silver/silver-chloride electrodes exhibit photosensitive characteristics. It is not fully understood whether the effect of light on the silver/silver-chloride electrode actually results in a potential being generated, or whether it results simply in a change in the electrode's offset potential. Nevertheless, in some instances, it may be desirable to shield the electrode from light interference to insure satisfactory operation. So far we have been unable to detect any photosensitive characteristics in the Tektronix silver/silver-chloride electrodes.
- silver
purity It should be noted that if silver is used for electrodes, either in its raw state or as a silver/silver-chloride electrode, it should be of spectroscopic grade (99.999% pure) to ensure noise-free operation. "Fine" jeweller's silver is nowhere near pure enough.

16.4 REUSABLE SURFACE ELECTRODES

reusable
versus
disposable

There are two principal categories of electrodes in use -- reusable electrodes and disposable electrodes. Reusable electrodes are intended to be used on many subjects after having been cleaned and perhaps sterilized after each use. Disposable electrodes are intended to be discarded after use on a single subject. Price is rarely a paramount factor in selecting a reusable electrode type; however, disposable electrodes must be inexpensive to be practical thereby limiting their performance. Reusable electrode types usually offer improved performance over disposable electrode types and may be expected to cost between one dollar and ten dollars for each electrode.

indirect
contact

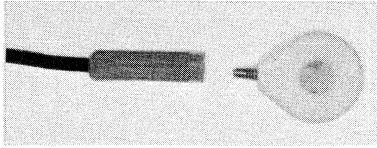
Each of the two principal electrode types (reusable and disposable electrodes) can be further categorized into indirect contact (floating) and direct contact types. Indirect contact electrodes are built in such a way as to be spaced some distance from the skin and rely on an electrolytic bridge between the electrode and the skin. This electrolytic bridge is formed by conductive electrode paste or "jelly," such as the electrode paste manufactured by Day-Baldwin Inc. that is currently supplied as a standard accessory with the Tektronix Type 410 Monitor. Indirect contact electrodes typically produce less motion artifact than the direct contact types and their performance is somewhat more predictable as the electrode is always used in conjunction with only one type of electrolyte -- the electrode paste.

direct
contact

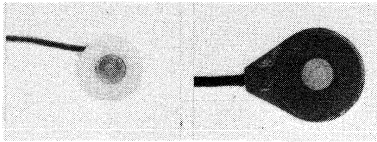
Direct contact electrodes, as the terminology implies, are designed to make direct contact with the subject and they may, if necessary, be applied without electrode paste (dry). Since most direct contact electrodes are used with a small amount of electrode paste to improve electrode-skin contact impedance, they should probably more correctly be categorized between the indirect contact and the direct contact types. The electrodes supplied by Tektronix for use with the Type 410 Physiological Monitor and also the Beckman electrodes shown in Fig. 16-9 are indirect-contact silver/silver-chloride electrodes. It is generally agreed that this type electrode provides the best available electrode performance for surface electrode applications.

indirect-
contact
best
performance

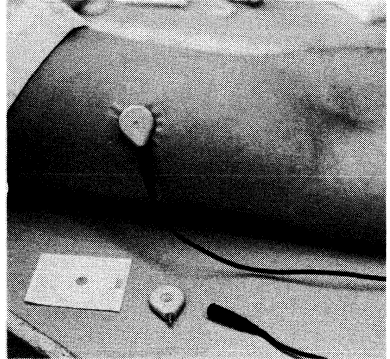
ELECTRODES SUPPLIED BY TEKTRONIX FOR USE WITH THE TYPE 410 MONITOR



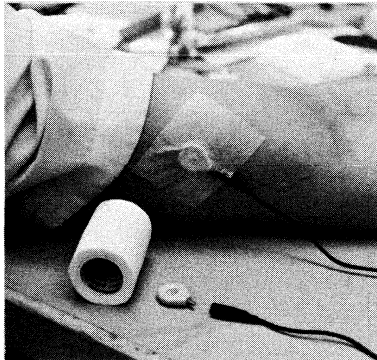
Type currently supplied



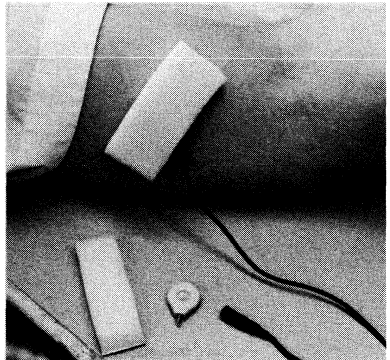
Earlier types supplied



Recommended application configuration

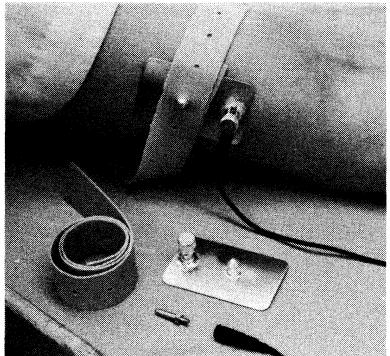
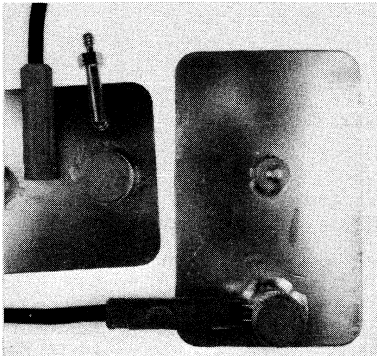


Alternate application configuration

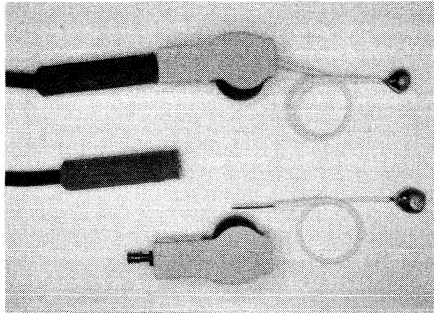


Alternate application configuration

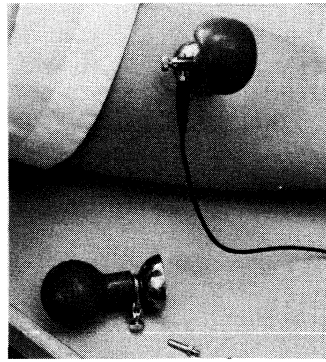
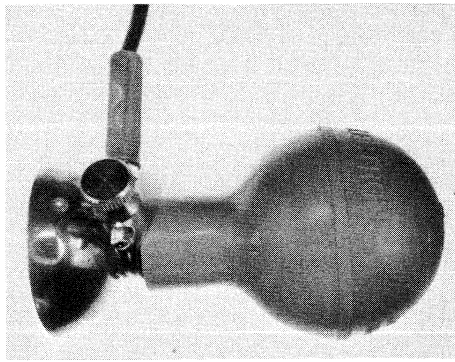
SILVER PLATE ELECTRODES



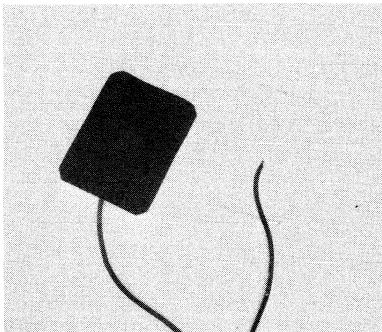
SILVER EEG ELECTRODES



SELF RETAINING ELECTRODES



LEAD PLATE GSR ELECTRODES



BECKMAN ELECTRODES

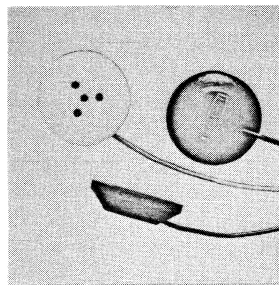


Fig. 16-9. Reusable electrode types.

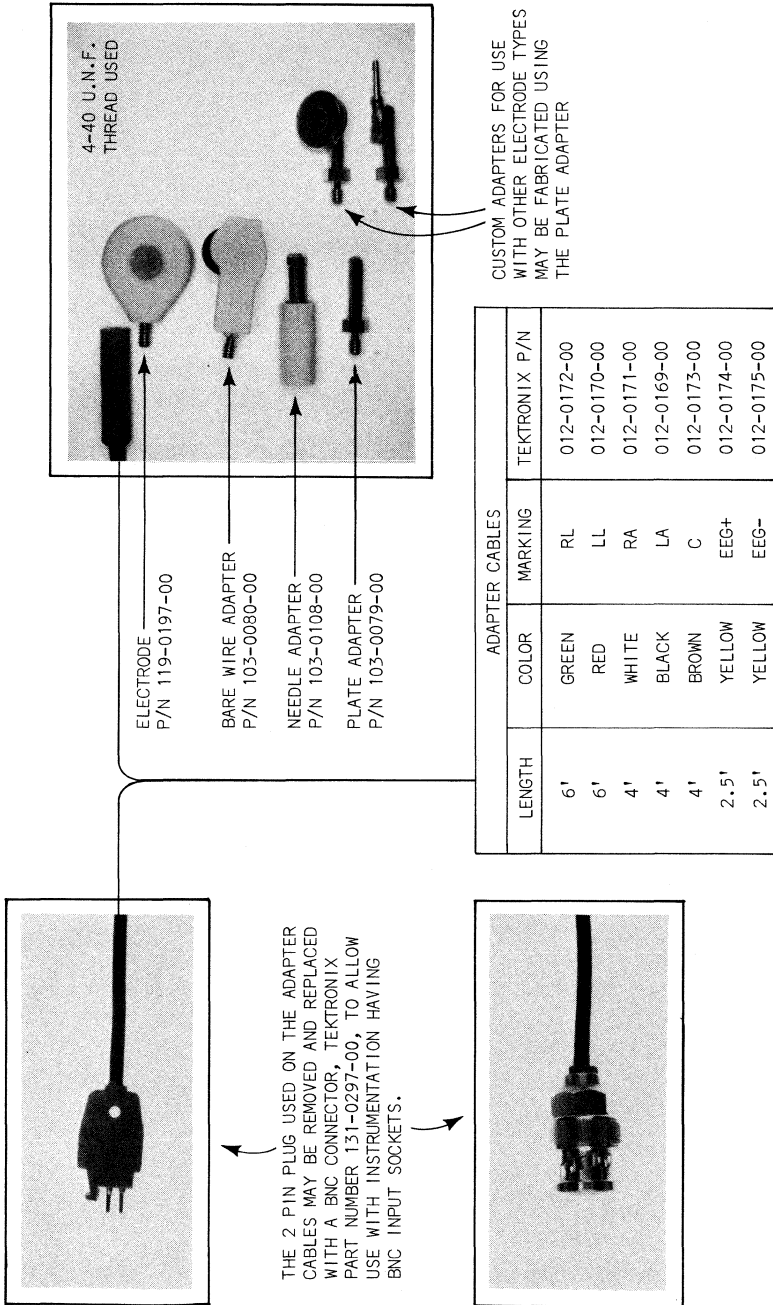


Fig. 16-10. Tektronix electrode cables and adapters.

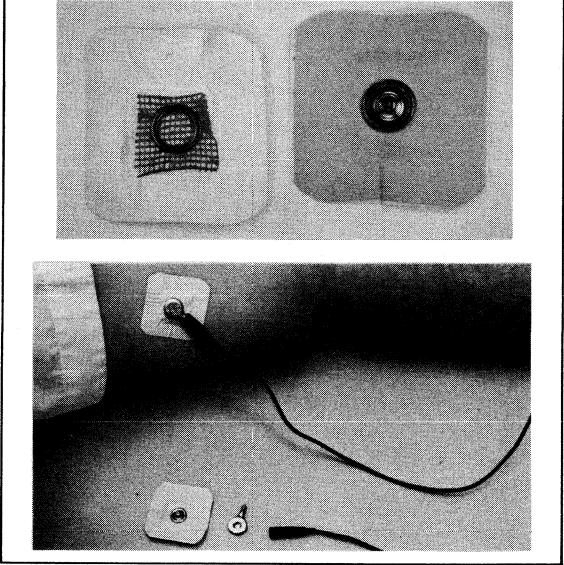
reusable
electrode
types

Fig. 16-9 shows various types of reusable surface electrodes. Earlier electrode types supplied by Tektronix for use with the Type 410 Physiological Monitor were permanently affixed to an electrode cable; however, the types currently supplied are detachable from the electrode cable. Fig. 16-9 shows the current Tektronix silver/silver-chloride electrode applied to a subject using the adhesive rings and electrode paste supplied by Tektronix with the Type 410 Monitor. Alternative application configurations, using surgical tape over the electrode or using an adhesive-backed foam pad over the electrode, are also shown. The large, silver-plated, copper electrodes historically used to record the ECG, the self-retaining electrodes currently preferred as a chest electrode for ECG recording, silver EEG electrodes, lead plate electrodes intended for GSR recording, and silver/silver-chloride electrodes produced by Beckman are shown. All of the electrodes in Fig. 16-9 are photographed to the same scale. These electrodes can be adapted for use with the Tektronix Type 410 Physiological Monitor using either a plate adapter, a needle adapter or a bare-wire adapter. These adapters are shown in use in Fig. 16-9 and further details on adapter cables and adapters available from Tektronix are shown in the next figure.

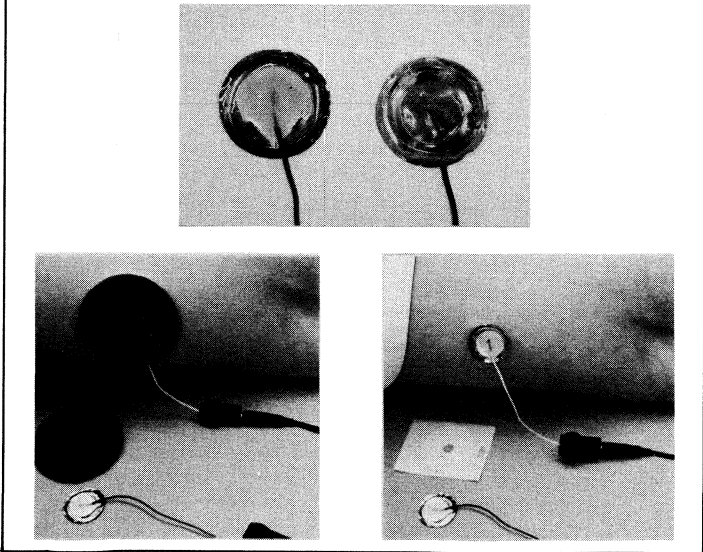
adapter
cables

Referring to Fig. 16-10, Tektronix supplies 2½-, 4- and 6- foot adapter cables to allow signals to be connected from electrodes to the Type 410 Monitor. These adapter cables can be directly connected to the Tektronix silver/silver-chloride electrodes or can be connected to other electrode types via either plate adapters, needle adapters or bare-wire adapters. These are the only adapter types available from Tektronix; however, enterprising users have successfully adapted other electrode types for use with the Tektronix Type 410 Monitor. The electrode and adapter system shown in Fig. 16-10 is primarily intended for use with the Tektronix Type 410 monitor; however, if the two-pin plug supplied with these cables is replaced by a BNC connector, these cables and adapters can be used with other instrumentation having BNC input sockets.

UNITED AIRCRAFT CORPORATION; HAMILTON STANDARD
DIVISION; ADHESIVE "TELECTRODE" DISPOSABLE ELECTRODE



ELECTRONICS FOR MEDICINE INC. DISPOSABLE PURE SILVER ELECTRODES



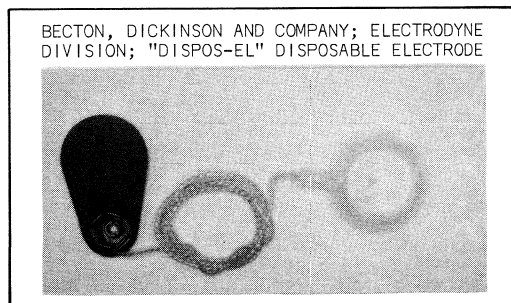
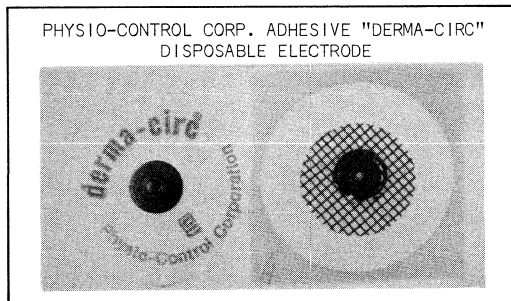
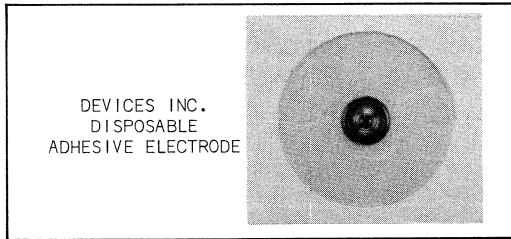
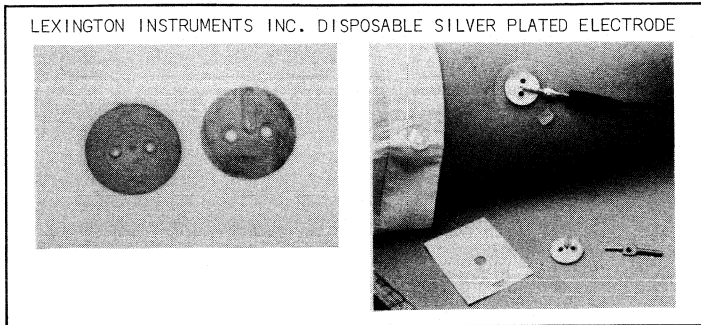


Fig. 16-11. Disposable electrode types.

16.5 DISPOSABLE SURFACE ELECTRODES

disposable electrode types

Tektronix does not produce a disposable electrode for use with the Type 410 Physiological Monitor. Disposable electrodes produced by six different manufacturers are shown in Fig. 16-11. As stated earlier, disposable electrodes must be inexpensive and one would expect to pay less than a dollar for each electrode. While disposable electrodes are indeed convenient and offer adequate performance for many applications, they appear to be unsuited to applications where the subject may receive stimulation and/or defibrillation as their electrode offset potential stored-charge characteristics appear to be substantially worse than for silver/silver-chloride electrodes. While Tektronix would not wish to recommend any particular brand of disposable electrode, all are usable with the Type 410 Monitor for ECG recording if the patient is not expected to require defibrillation, stimulation or cauterization.

spray-on electrode

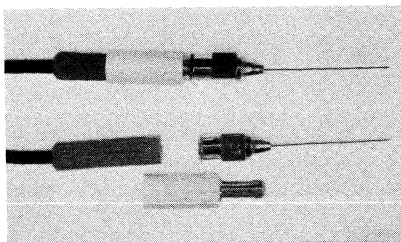
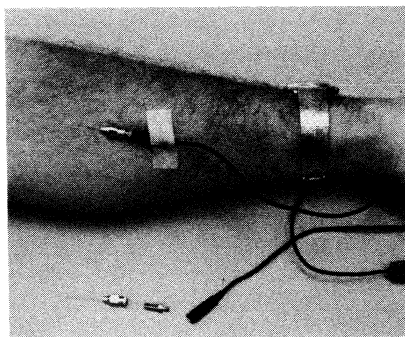
A relatively new and unique disposable electrode is the spray-on electrode developed by NASA for ECG recording in space flight. This electrode system is commercially available from the Hauser Research and Engineering Company in Boulder, Colorado. These electrodes basically consist of a film of conductive adhesive that is sprayed onto the skin with a contact wire imbedded into this adhesive during its drying cycle. These spray-on electrodes are particularly suited for long-term monitoring on subjects who are physically active.

16.6 NEEDLE ELECTRODES

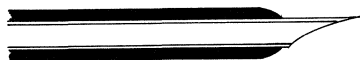
hypodermic electrodes

Needle electrodes fall in the disposable electrode category and are primarily used for ECG monitoring during surgery or where extremely fast electrode application is desirable, as in an emergency situation. Needle electrodes may simply be stainless steel hypodermic needles. Stainless steel is a somewhat unsatisfactory material for electrodes (covered elsewhere in this chapter) as its

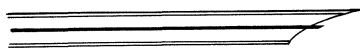
characteristics are inferior to most other electrode materials. Stainless steel must however, be used as it is one of the few materials that can offer sufficient mechanical strength. Typical long-needle electrodes are shown in Fig. 16-12. These needles are regular hypodermic needles but *short* hypodermic needles are preferable as it is only necessary to insert the needle just under the subject's skin.



(A) HYPODERMIC NEEDLE ELECTRODES



INSULATED HYPODERMIC NEEDLE
(OR SURGICAL NEEDLE)



CONCENTRIC NEEDLE ELECTRODE
(HYPODERMIC NEEDLE WITH ONE
CONDUCTOR THREADED THROUGH IT)



DIFFERENTIAL CONCENTRIC NEEDLE
ELECTRODE (HYPODERMIC NEEDLE
WITH TWO CONDUCTORS)

(B) CONCENTRIC NEEDLE ELECTRODES

Fig. 16-12. Needle electrodes (disposable).

The hypodermic needle electrode referred to previously is intended to perform similar functions to those of the surface electrodes. Special types of needle electrodes, such as concentric needle electrodes or insulated needle electrodes, may be used in applications where it is necessary to insert the needle to a bioelectric site *within* the body rather than to record the electrical activity produced near the surface of the body. Concentric needle electrodes may be two or three inches long for use in electromyography of the skeletal muscles and may be up to six inches long for certain applications in neurosurgery. Various arrangements of concentric needle electrodes are also shown in Fig. 16-12B. The simplest form of "concentric" needle electrode is an *insulated* hypodermic needle. The insulated needle effectively only records electrical activity generated in the region of the tip of the needle. A simple concentric needle electrode consists of a hypodermic needle with a length of 36-gauge enamel-covered silver wire threaded through it. The enamel covering on the silver wire insulates it from the stainless steel needle and is removed at the electrode tip area to provide a needle electrode similar to a miniature coaxial cable. A bioelectric signal may be recorded by connecting the recording amplifier between the center conductor and the outer concentric needle or, preferably, two concentric needle electrodes may be used with a recording amplifier connected between the center conductor of each and the outer needle of each acting as a grounded shield. Two pieces of enamel-covered platinum wire may be used in a hypodermic needle to produce a differential concentric needle electrode. This form of electrode is primarily used with the outer needle grounded and with the recording equipment connected between the two inner conductors. This effectively records the electrical activity produced in the region of electrodes spaced less than 1 millimeter apart. This type of differential concentric needle electrode may be used for recording the EMG and evoked responses.

concentric
electrode

differential
concentric
electrodes

16.7 MICROELECTRODES

Microelectrodes are used to record the electrical activity within individual cells, thus, the active electrode tips of these microelectrodes must be small enough so as not to interfere with the cell's normal function. Microelectrodes may be expected to have tip diameters within the range from 0.1 micron to 10 microns (one micron = 10^{-6} meters).

glass
micro-
electrodes

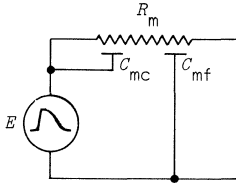
Microelectrodes can be divided into two broad classifications -- glass microelectrodes and metal microelectrodes. Glass microelectrodes consist of an electrolyte-filled glass tube that has been drawn to a capillary with a silver wire making connection to the electrolyte within the tube. The transfer from ionic conduction to electronic conduction is made at the relatively large surface area interface between the platinum wire and the electrolyte. Metal microelectrodes consist of a finely tapered metallic rod covered with a layer of insulation. With metal microelectrodes, the transfer from ionic conduction to electronic conduction occurs at the electrode tip, that is, within the cell itself.

metal
micro-
electrodes

The characteristics of glass microelectrodes are covered in considerable detail in Chapter 11. The equivalent electrical circuit of the glass microelectrode is shown in Chapter 11 and is also shown in Fig. 16-13 together with the equivalent circuit for the metal microelectrode. Referring to the equivalent circuit for the metal microelectrode, most of the microelectrode resistance (R_m) is located within the first few microns from the electrode tip. With the metal microelectrode inserted into the cell, this resistive area will be located within the cell. The capacity between the electrolyte within the cell and the metal microelectrode (with the microelectrode insulating material acting as the dielectric) is represented by C_{mc} .

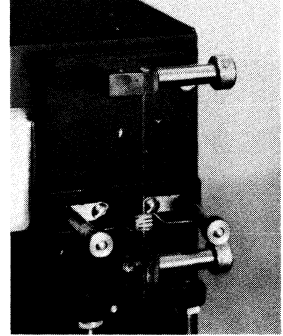
micro-
electrode
character-
istics

GLASS MICROELECTRODE WITH SILVER WIRE

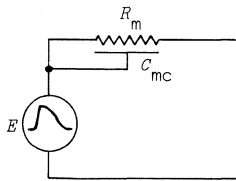
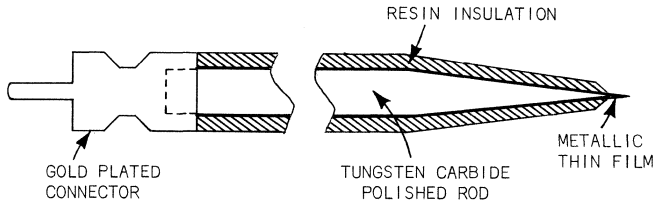


TYPICAL:
 $C_{mc} = 0.1 \text{ pF}$
 $C_{mf} = 0.2 \text{ pF}$
 $R_m = 10^8 \Omega$

A GLASS "MICROELECTRODE PULLER"



RESIN INSULATED METAL MICROELECTRODE



TYPICAL:
 $C_{mc} = 15 \text{ pF}$
 $R_m = 10^6 \Omega$

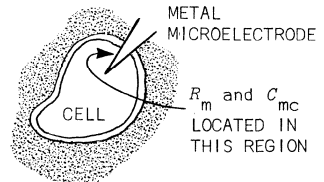


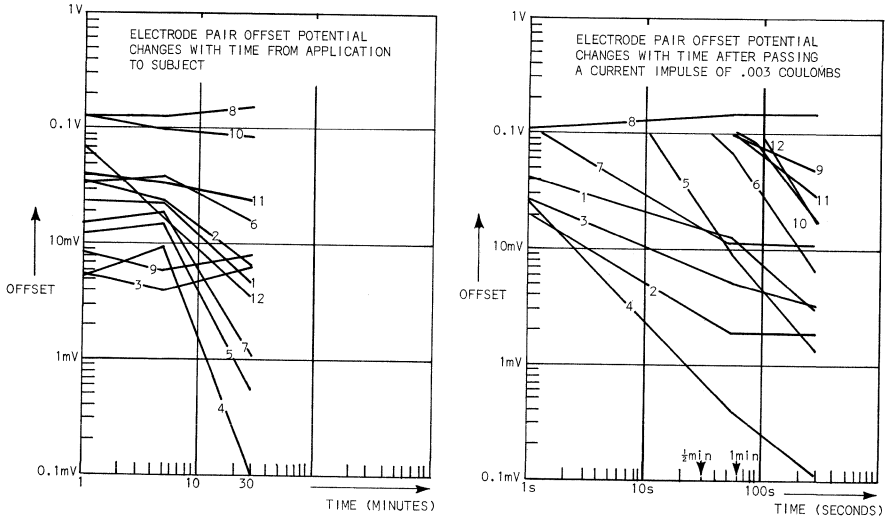
Fig. 16-13. Microelectrodes.

micro-
electrode
construction

Metal microelectrodes are formed by etching a metal rod while the metal rod is being slowly withdrawn from the etching solution, thus forming a tapered tip on the rod. The etched metal rod is then dipped into an insulating solution bath to form a layer of insulation over the rod. This insulating layer is prevented from covering the tip by the surface tension action. Glass microelectrodes are formed by heating a special quality glass tubing and then drawing the soft glass tubing apart with controlled tension to form a capillary. This is usually achieved with a mechanical device known as a "microelectrode puller" consisting of a heating element and tensioning system. A microelectrode puller is shown in Fig. 16-13.

16.8 ELECTRODE APPLICATION

Electrodes may be affixed to a subject in a variety of ways and application recommendations are usually included with electrodes at the time of purchase. Fig. 16-9 shows various techniques for applying many electrode types and, in the case of the Tektronix electrodes, shows application by either an adhesive ring, adhesive tape or an adhesive foam pad. The silver EEG electrode shown in Fig. 16-9 is glued to the subject's forehead with a commercial adhesive known as collodion. All electrode types offer better performance if applied after cleaning and abrading the skin with a small amount of slightly abrasive electrode paste. This tends to remove dead skin cells from the surface and thus decreases the electrode contact impedance.



ref	REUSABLE ELECTRODES - SEE FIG. 16-9	RESISTANCE AFTER 40 min
1	CURRENT TEKTRONIX ELECTRODES	170Ω
2	EARLIER TYPE TEKTRONIX ELECTRODES - SMALL	320Ω
3	EARLIER TYPE TEKTRONIX ELECTRODES - LARGE	100Ω
4	SILVER PLATE ELECTRODES	260Ω
5	SELF RETAINING ELECTRODES	250Ω
6	"SOLDER" EEG ELECTRODES	400Ω
7	LEAD PLATE GSR ELECTRODES	260Ω
	<u>DISPOSABLE ELECTRODES - SEE FIG. 16-11</u>	
8	UNITED AIRCRAFT	380Ω
9	ELECTRONICS FOR MEDICINE	360Ω
10	LEXINGTON INSTRUMENTS	280Ω
11	PHYSIO-CONTROL CORP	500Ω
12	NEEDLE ELECTRODES - SEE FIG. 16-12	350Ω

NOISE AFTER 30 min
50μV/DIV

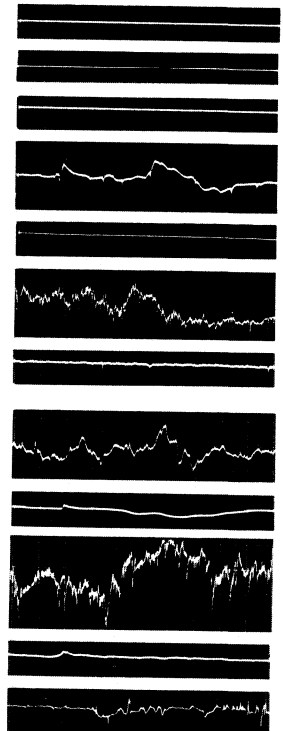


Fig. 16-14. Characteristics of electrode types shown in Figs. 16-9, 16-11 and 16-12.

16.9 STIMULATING ELECTRODES

stimulating
versus
recording
electrodes

Almost all of the electrode types discussed previously may be used as stimulating electrodes. When using electrodes for stimulation, current is deliberately passed between the electrodes. This current, unless it is truly biphasic, will increase the electrode offset potential, thus making the electrodes increasingly unsuitable for future use as recording electrodes, even if silver/silver-chloride electrodes are used as stimulating electrodes. It is advisable that, once any electrode has been used as a stimulating electrode, it be identified in some manner and not used thereafter as a recording electrode. Silver should be used for stimulating electrodes as it is nontoxic and does not tend to produce iontophoresis due to the passage of electric current. Iontophoresis is the driving of external ions in through the skin which can produce considerable irritation if unsuitable electrodes are used.

16.10 COMPARISON OF ELECTRODE TYPES

30-minute
drift
comparison

There is undisputed evidence to suggest that silver/silver-chloride electrodes are far better than other types for recording physiological signals; however, they are not well suited to some applications and may offer unnecessary performance in other applications. Many non-silver/silver-chloride electrodes are used for physiological monitoring and are discussed in previous sections in this chapter. A comparison between 12 of the electrode types discussed in this chapter is shown in Fig. 16-14. Several pairs of each of the 12 electrode types were tested and, in each case, the pair exhibiting the poorest performance characteristics is shown in this characteristic summary. The first graph represents the electrode offset potential drift of each of these 12 electrode types over a 30-minute period after the electrodes were applied to a subject and coupled to an amplifier having a 2-megohm differential-input impedance. In all cases, the reusable electrodes had been used several times previously.

applying
current
pulse

recovery
comparison

noise
comparison

conclusion

After observing the drift characteristics of these electrodes over a 30-minute period, a current impulse of .003 coulombs was applied through the electrodes by discharging a capacitor across them. The second graph shows the change in electrode offset potential for each of these 12 electrode types during the first five minutes after application of the current impulse. In all cases the electrodes were tested in an electrolyte consisting of 0.9 percent sodium chloride. The insertion of the electrodes into the sodium chloride bath was adjusted to match the electrode surface area in the bath to the electrode surface area normally in contact with a subject. The separation of the electrodes within the bath was maintained at approximately three centimeters. Fig. 16-14 shows the resistance of the electrode pair to a 10 mA DC current after having been in the electrolyte and connected to the amplifier for a period of 40 minutes. Fig. 16-14 also shows the noise produced by these electrodes over a 10-second period after having been in the electrolyte bath and connected to the amplifier for a 30-minute period.

While we would not wish to make a detailed analysis of the results shown in Fig. 16-14, it is apparent that reusable electrode types generally exhibit lower electrode offset potential and lower noise characteristics than the disposable electrode types. It should also be noted that, particularly for disposable electrodes, manufacturing techniques are constantly changing and the performance typical of an electrode type may change as the manufacturing process is changed.

17

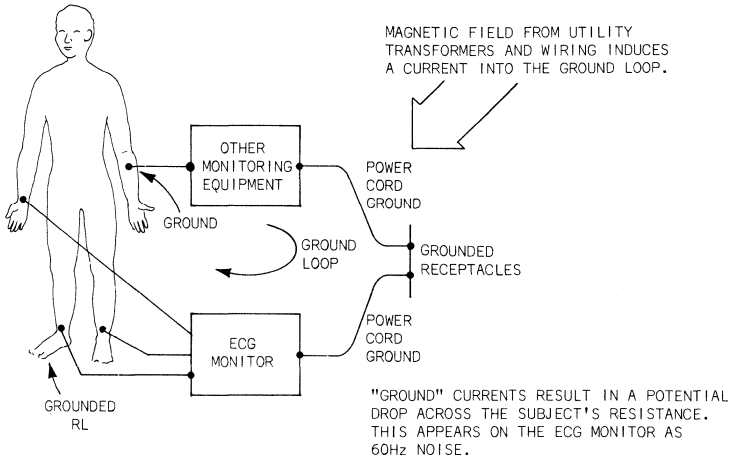
GROUNDING - SAFETY

artifact

The term artifact refers to the presence of an unwanted signal such as power line frequency interference or noise. When recording, for example, the EEG, the presence of ECG or EMG information on the recording also constitutes artifact. By far the most common source of artifact is power line interference; however, power line leakage considerations are of paramount importance from a safety viewpoint and it is not uncommon for safety considerations to seemingly contradict artifact-elimination considerations.

17.1 GROUNDING

If a subject were deliberately connected across a one-volt AC source, an AC current of perhaps 100 μ A would flow through the subject due to the finite impedance ($\approx 10,000$ ohms) of the subject and skin connections. Although such a connection should never be deliberate, many instrumentation systems inadvertently produce a similar situation in the form of a ground loop.



IF THE GROUND LOOP CANNOT BE ELIMINATED BY "OPEN CIRCUITING" THE LOOP BY REMOVING A GROUND CONNECTOR; EITHER

INCREASE IMPEDANCE OF GROUND LOOP SO SUBJECT IMPEDANCE IS NEGLIGIBLE (ADD 100kΩ IN SERIES WITH RL)

OR

REWIRE GROUNDS SO THAT GROUND CURRENTS DO NOT PASS THROUGH SUBJECT (USE SINGLE POINT GROUNDING).

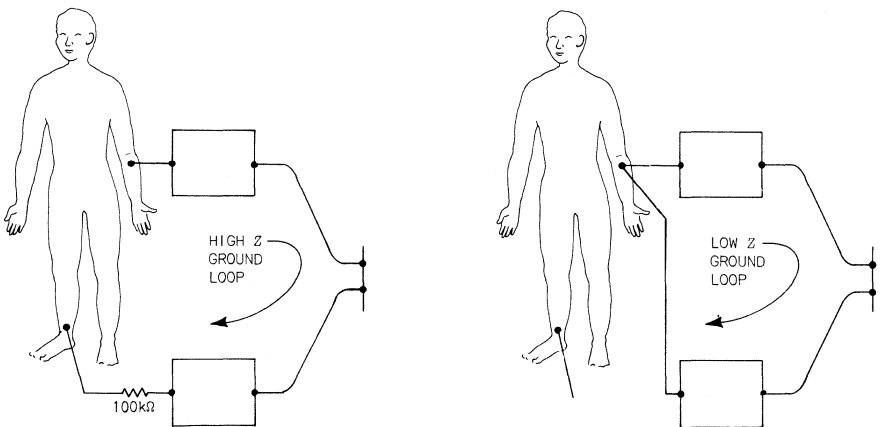


Fig. 17-1. Induced ground current from a "ground loop".

induced
60-Hz

If a 10-foot length of wire is connected between a patient's left arm and his right leg, a 60-Hz AC current will be induced through the patient as this wire acts as a transformer secondary, with current magnetically induced from adjacent power wiring and transformers. Although the magnetic coupling material involved is air and the amount of coupling therefore is extremely small, it is *finite* and an EMF will be induced into the loop which will in turn cause current to flow through the subject. This 10-foot length of wire may, in practice, be two separate grounding wires attached to a subject (Fig. 17-1).

ground
loop

Referring to Fig. 17-1, if either the ECG monitor or the "other monitoring equipment" is connected to the subject, a ground loop is not produced. However, if both instruments are simultaneously connected to the subject, the ground connections for each instrument form a ground loop. The ground loop shown produces a potential between the subject's right leg and left arm causing a current to flow through the patient's bulk body resistance. This current flow produces a voltage drop across different parts of the body which would appear as a 60-Hz artifact on the ECG monitor.

eliminating
ground
loop

The ground loop produced by the configuration shown in Fig. 17-1 may simply be eliminated by removing one of the ground connections to the subject whenever both items of monitoring equipment are used simultaneously. If this is particularly inconvenient, the ground loop may be eliminated by adding a high impedance in series with one of these ground leads to reduce current flow to a negligible value or the subject may be grounded at a single reference point so any induced current flow does not flow through the subject's bulk body impedance. Both of these ground loop changes are shown in Fig. 17-1 and both changes effectively eliminate the current flow through the subject, thus eliminating 60-Hz potential differences between various points on the subject.

17.2 INDUCED GROUND CURRENTS

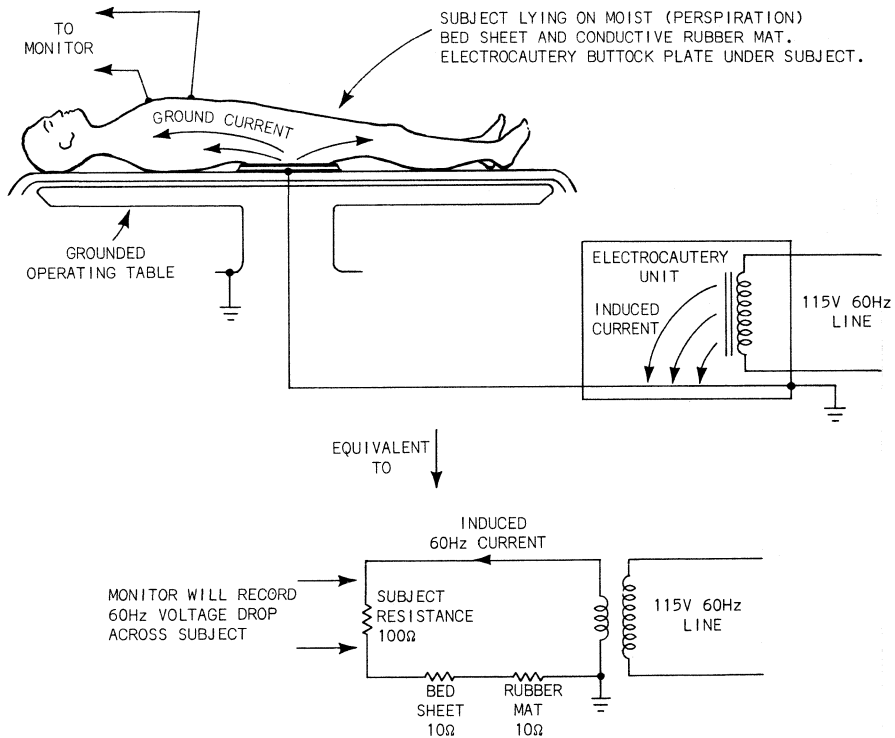
transformer
induced
current

A ground loop is an example of an induced ground current, the value of which can be reduced or eliminated by modifying the grounding configuration. Another form of induced ground current is illustrated in Fig. 17-2 where a current is induced into a ground wire via an instrument's power transformer. In this specific case, an induced 60-Hz ground current is induced into the grounding circuit of an electrocautery unit. This ground current flows via the electrocautery ground plate, through the subject and to ground via a perspiration-dampened bedsheet, a conductive rubber mat and the grounded operating table.

buttock
plate
ground
loop

old style
electro-
cautery
units

The above type of induced ground current is particularly difficult to eliminate as safety considerations demand that the subject be grounded and that the electrocautery-unit buttock plate be well grounded to the electrocautery unit and to the subject to eliminate burns at electrode sites during electrocautery. This particular problem is often encountered when trying to monitor a subject's ECG in an operating room using an older style electrocautery unit. When these older units were designed it was not expected that the ECG may be monitored while the electrocautery unit was connected to the line ready for use. Modern electrocautery units have more carefully designed internal-grounding configurations and incorporate shielding to prevent any magnetically induced or capacitively induced current from entering the subject via the buttock plate. Perhaps the only solution that can be offered for an electrocautery unit of the older design is to discard the unit. Although this "solution" may seem harsh, the laws of physics offer no alternative and, more importantly as will be seen later in this chapter, the induced current produced by the electrocautery unit may be sufficient to cause death if grounded catheters are used on the subject.



AN INDUCED 60Hz CURRENT OF 350nA RMS WILL PRODUCE
2mm OF 60Hz NOISE ON A STANDARD ECG MONITOR

Fig. 17-2. Induced ground current from electrocautery unit.

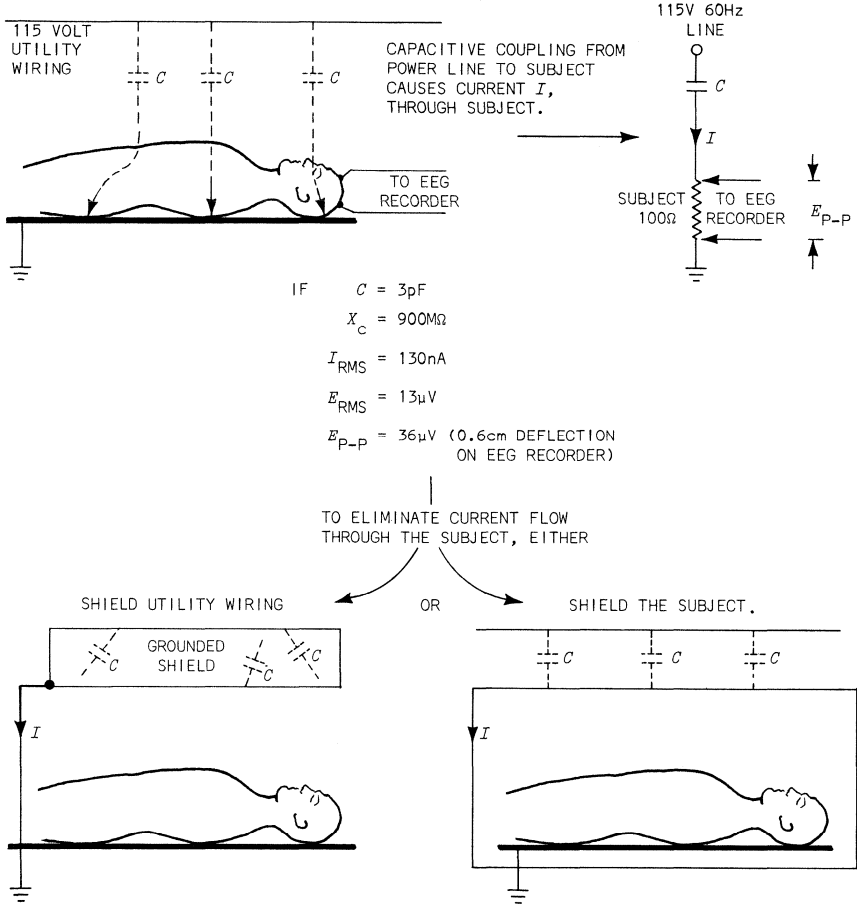


Fig. 17-3. Electrostatic shielding.

17.3 ELECTROSTATICALLY INDUCED CURRENTS

utility
wiring
capacitance

Capacitive coupling between the 115-volt wiring and the subject constitutes an impedance which will allow current to flow through the subject. The situation is illustrated in Fig. 17-3 which shows that a stray capacitance of 3 pF between the utility wiring and the subject can produce 6 centimeters of 60-Hz artifact on an EEG recording.

utility
wiring
shielding

To eliminate electrostatically induced currents, it is necessary to provide an alternative path for this current to flow. This may be achieved by either shielding the 115-volt utility wiring, in which case the current flows from this wiring to the shield, or by placing the subject inside an electrostatic shield (Faraday Cage), in which case the current flows to ground via the shield rather than via the subject. In either case, shielding is expensive and it may be preferable to locate a recording site in an alternative part of the hospital or laboratory where electrostatically induced current is less bothersome. In the research environment, where high sensitivity amplifiers are invariably used, the subject and parts of the recording equipment should be enclosed in a screened room to effectively isolate the subject from capacitively induced currents from any source.

recording
lead
shielding

Currents may be capacitively coupled to the leads connecting the subject to the recording equipment, thus these leads must be screened to effectively shield them. Care should be taken to ensure that these shields are, however, only grounded at one point otherwise, in attempting to eliminate interference by electrostatic shielding, one may inadvertently create 60-Hz interference by introducing a ground loop. This interference may be produced by 60-Hz power wiring or it may be high-frequency interference produced by radio and/or television transmitters.

17.4 ELECTRIC SHOCK CURRENT THRESHOLDS

When electric current is passed through the body, part of this current will pass through the heart and interfere with the normal function of the heart, perhaps causing death. In cases of death by electrocution, almost all subjects are killed by the passage of electric current through the heart rather than by some related occurrence such as burns or

voltage
versus
current
"shock"

muscular paralysis of the muscles controlling breathing action. If a person receives a 115-V shock, it is not the voltage of this shock that is important but rather the *current* through the body. Thus, if a person receives a 115-V "shock" standing in a dry environment insulated by normal clothing, the "shock" may hardly be felt as the current may be well below 1 mA. If, however, a person receives a 115-V shock while standing barefooted on moist ground, the person would receive a severe shock and probably be killed as the current may exceed 100 mA.

shock
current
thresholds

Referring to Fig. 17-4; experimental investigation on numerous subjects (both male and female) has indicated that up to 300 μ A applied to the surface of the body, such as from one arm to the other, may be suggested as being reasonably safe for most subjects. It can be seen that 99.5 percent of the population require 400 μ A or more of 60-Hz current to perceive the current and that the threshold of perception or sensation increases as frequency increases above 100 Hz and below 10 Hz. At 10,000 Hz and at DC, the threshold of sensation is approximately five times greater than at 60 Hz. While current above the threshold of sensation may not be detrimental to a healthy person, it may cause complications in a hospitalized patient and will certainly create anxiety which would be detrimental to the patient's general well being. From Fig. 17-4 it can also be seen that "cannot-let-go" current thresholds are an order of magnitude above sensation current thresholds and that they also increase for frequencies below 10 Hz and above 100 Hz.

ventricular
fibrillation

vulnerable
period in
cardiac
cycle

Fig. 17-5 shows the physiological effects of 60-Hz arm-to-arm current and shows a threshold of sensation at 300 μ A, a threshold of pain at 1 mA, a "cannot-let-go" threshold at 10 mA and a ventricular fibrillation-induction threshold at 100 mA. 99.5% of the population will have thresholds above these values. Ventricular fibrillation refers to malfunctioning of the ventricular musculature which will interfere with the normal blood-pumping action of the heart and eventually cause death. Ventricular fibrillation is produced by current directly through the heart during a specific portion of the cardiac cycle known as the "vulnerable period." The vulnerable period for ventricular muscle occurs during the upstroke of the T-wave and a single shock impulse lasting for less than 0.1 second could cause ventricular fibrillation if received during this vulnerable period.

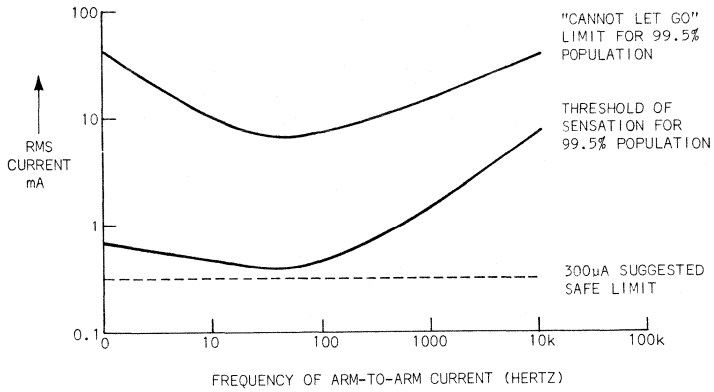


Fig. 17-4. Physiological effects versus current frequency.

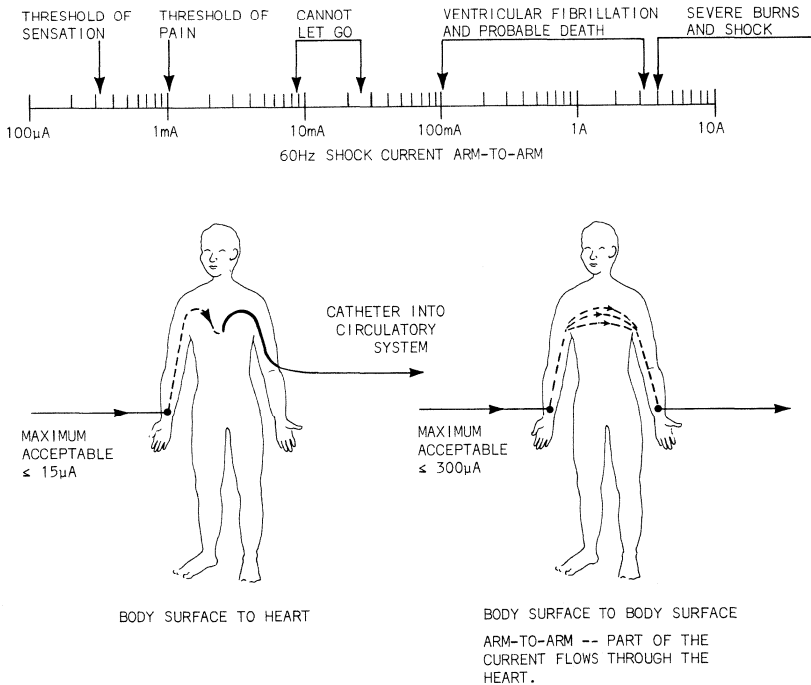


Fig. 17-5. Physiological effects of 60Hz electric current.

- current density versus thresholds
- The 300 μA threshold refers to 60-Hz current applied from one arm to the other arm and is intended as a guide to acceptable leakage current levels. The acceptable level for any one subject may be somewhat above this value and will also depend on the size of the current contact. A very small point contact can undoubtedly be felt at 0.3 mA, but a current in excess of perhaps 1 mA may not produce sensation if the contacts are somewhat larger. Depending on the size of contact, the threshold of pain may also be considerably above 1 mA, probably 10 mA if the contacts are large enough. It is thus evident that current density is also important in determining the physiological effects of electric current.
- internal current thresholds
- The threshold of sensation of electric currents differs greatly between currents applied arm-to-arm and currents applied internally to the body. If a current is applied internally, a far greater percentage of the current may flow via the arterial system directly through the heart, thus less over-all current is required to produce ventricular fibrillation. Experiments with dogs have indicated that in some animals ventricular fibrillation can be produced with currents as low as 17 μA applied directly to the dog's heart. In general, as the threshold of sensation in man may not be much higher than for the dog, 30 μA of 60-Hz current through the heart may produce ventricular fibrillation and 15 μA could perhaps be considered as a safe upper limit. 5 μA has been postulated as a safe upper limit under certain adverse conditions.
- internal probes and electrodes
- A probe or electrode within the body such as a cardiac catheter or pacemaker electrode will provide a direct electrical pathway to the heart and thus the 15 μA current limit proposed in the preceding paragraph should be considered as an upper limit for current flow through these conductors. The impedance from these conductors to the subject's skin surface may typically be 1,000 ohms, thus a potential of 30 mV between these conductors and a point on the surface of the subject's body is sufficient to cause electric shock, ventricular fibrillation and possible death! It has been proposed that the introduction of the *internal electrode* has been the leading factor in the present high incidence of accidents in hospital patient care areas and operating rooms.
- high accident rate

Cardiac catheterization for diagnostic or therapeutic purposes has become common in recent years. Sensing catheters may have electrodes or transducers at their tips for ECG recording, blood pressure measurements and other diagnostic procedures. Fluid-filled catheters are also in common use and, although they are not normally considered to be a probe or electrode within the body, the fluid in these catheters may be conductive, thus they also may provide an electrical pathway to the heart.

17.5 INSTRUMENTATION SAFETY CONSIDERATIONS

Twenty years ago the electrocautery unit was the only item of electronic equipment routinely used in the operating room. Any electrical shock hazard associated with this instrument was far outshadowed by the risk of explosion from flammable anesthetic gases. Nowadays the operating room is packed with electronic equipment, all of which is necessary for the fulfillment of today's modern surgical procedures, and the risk of explosion from flammable anesthetic agents is considered of secondary importance. All of the instruments used in the operating room may be used alone with relative impunity however, in combination, they produce an unprecedented electrical shock hazard of giant proportions. The interaction of various units connected to the subject demand a *system* approach which may be foreign to the training of medical staff and may also not be fully understood by the manufacturers of the various items of instrumentation. Some medical instrument manufacturers tend to be oblivious to this problem and tend to regard their product as an entity within itself rather than as a part of an over-all system.

The major source of potentially lethal currents in any instrument is leakage current from the 115-V (or 230-V) power-transformer primary. This leakage current is largely due to capacitive coupling from the power transformer primary to other parts of the transformer and/or other parts of the instrument. Instruments are usually designed so this leakage current flows to the instrument case and then to ground via the three-wire power cord provided with the instrument. Most modern general-purpose instrumentation may be expected to produce up to 500 μ A of leakage current, however instrumentation

unprecedented
shock hazard
demands
system
approach

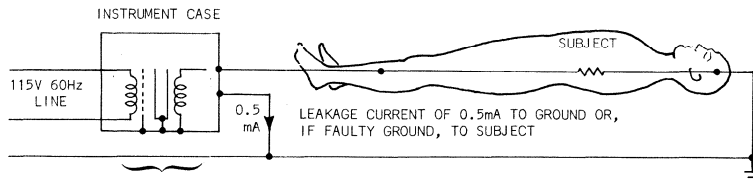
power
transformer
leakage
currents

shock
currents
from
ungrounded
instruments

designed for use in the medical environment should preferably have a leakage below $100 \mu\text{A}$. By special design techniques it is possible to decrease this leakage to well below $10 \mu\text{A}$. Although this may not always be practical, as long as the instrument is adequately grounded, the leakage current will not flow through the subject. A problem arises when instrumentation is used in an ungrounded configuration, as would be the case if the three-pin power plug provided with the instrument was replaced by a two-pin plug to allow the instrument to be used with the older style two-pin power receptacles. Under these conditions the ground leakage current may flow through the subject as shown in Fig. 17-6 and if one of the conductors is an internal electrode, such as a catheter, death will almost certainly result.

shock
currents to
isolated
subjects

It is apparent that if the subject were "floating" or isolated, then a path to ground would not be provided for the leakage current mentioned in the previous paragraph and no current would flow through the subject. In practice, however, this does not reduce the risk as fault currents could still flow between various items of equipment without involving "ground" in the current circuit; also, if a subject inadvertently touched a grounded item, such as a bed frame or water pipe, then the isolation would be nullified and electrocution may result. It is, therefore, apparent that protection systems must be organized on a grounded system concept.



CAPACITANCE FROM PRIMARY TO EITHER SHIELD, CORE OR GROUNDING SECONDARY RESULTS IN CURRENT FLOW FROM PRIMARY TO INSTRUMENT GROUND (CASE).

ASSUMING CAPACITANCE OF $.01 \mu\text{F}$:

X_C AT $60 \text{ Hz} = 250 \text{ k}\Omega$

$\text{RMS CURRENT} = \frac{115}{250} \text{ mA} = 0.5 \text{ mA}$

Fig. 17-6. Transformer capacitive leakage currents.

second
order
shock
protection

The various hazards mentioned so far in this chapter primarily refer to current flowing through the subject if a fault occurs or if something abnormal happens. As in any situation, faults do occur and abnormalities do happen, thus it is desirable to offer some second order protection to the subject.

current
limiting

Under normal operation of, for example, an ECG unit, only a small signal current should flow in any of the active electrodes or the right-leg ground electrode. Under abnormal or fault conditions, however, current may flow through any one of these paths and second order protection can be offered by incorporating some form of current limiting. The simplest form of current limiting, and a form adequate for use with the right-leg ground electrode, is to use a large value resistor in series with the electrode. Series resistance current limiting is not, however, always possible. In any amplification equipment, such as an ECG monitor, series resistance will degrade the common-mode rejection-ratio specification of the instrument to a point where 60-Hz interference may be intolerable. It is sometimes necessary to use a more sophisticated form of current limiting.

current
limiting
with field
effect
diodes

The Tektronix Type 410 Physiological Monitor is specifically designed for operating room use and incorporates series-resistance current limiting in the right-leg ground electrode and more sophisticated current limiting in series with the active electrodes. This current limiting utilizes field effect diodes, the impedance of which is approximately 1,000 ohms when the current is less than 1 μ A. However, the diodes limit fault currents to a maximum of 300 μ A as their impedance increases as the current attempts to exceed the 300 μ A level. It is thus apparent that current limiting within the Tektronix Type 410 Physiological Monitor provides adequate second order subject protection for "arm-to-arm" fault currents delivered by surface electrodes. It does *not* provide adequate protection to allow the Type 410 to be used with intracardiac-catheter electrodes. It should be noted that the instruction manual provided with the Tektronix Type 410 Physiological Monitor specifically states that the unit is *not intended* for use with internal electrodes of any type.

Many oscilloscopes produced by Tektronix and electronic instrumentation produced by other manufacturers are used in operating rooms and in intensive-care rooms on live human subjects. It should be noted that this type of instrumentation was designed for use in the electronics laboratory and does *not* incorporate any secondary subject protection. If a Tektronix product, other than the Type 410 Physiological Monitor, is to be used in conjunction with a human subject, we suggest that some form of current limiting be included in series with any electrode in contact with the subject. Spacelabs, Inc., Chatsworth, California, market an electrical shock protector which can be used in series with electrodes if the nominal line voltage is below 120-V AC or, alternatively, a current limiting adapter is described in Chapter 29 which is suited for use in areas with either 120-V or 240-V power. It provides secondary protection by limiting current to 300 μ A. Secondary protection does not guarantee safety, it simply reduces the probability of an accident. Each protection feature added to instrumentation should reduce this probability.

current
limiting
adapters

17.6 ELECTRICAL SERVICE GROUNDING

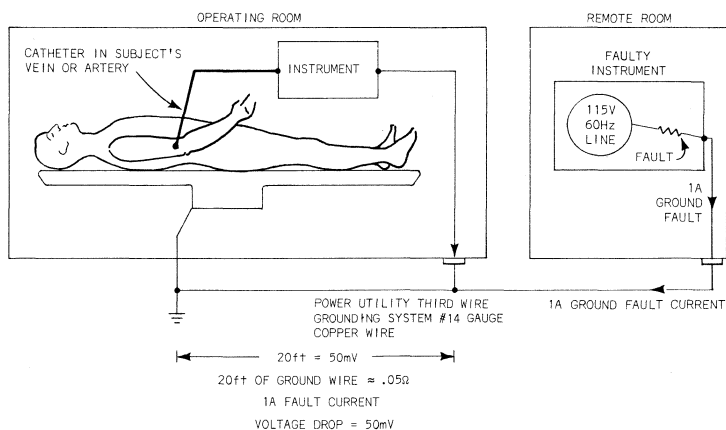
Section 17.4 pointed out that a potential of only 30 mV between two supposedly grounded conductors may allow enough current to flow through a subject to cause death by electrocution. If multiple ground points are available in an operating or intensive-care room there is no guarantee that all of these are at the same potential and a voltage may exist between, for example, a grounded wall outlet and a nearby water pipe. These subtle ground potential differences are difficult to detect, but they may be *fatal* if left undetected. It is not uncommon to find two electrical power services available in older hospitals, the second service having been provided to handle increased power demand since the hospital was constructed. This second service may come from a different part of the hospital and the potential at its ground conductor may differ from the potential of the ground conductor of the older service by a

potentials
between
different
grounding
systems

volt or more. If subtle differences in ground potential are detected, they can usually be almost completely eliminated by interconnecting each ground point with a heavy-gauge copper grounding bus. As currents of several amperes may flow in this interconnecting grounding bus, the copper wire used should be at least a 1/4 inch in diameter. Such subtle ground differences can be detected with a differential AC voltmeter or a differential oscilloscope.

Referring to Fig. 17-7, it is apparent that two ground points from the same power service may show a potential gradient due to a fault current in the power service. Although the fault current of 1 A shown in Fig. 17-7 is insufficient to trip a circuit breaker in the active line of the power service, it is sufficient to produce a voltage drop of 50 mV across 20 feet of standard service ground wire. Once again, this 50 mV may be enough to cause a fatality. To avoid this difficulty the service outlets should be grouped together to provide a single point ground.

remote
fault
currents



THIS 50mV APPEARS BETWEEN THE SUBJECT'S CIRCULATORY SYSTEM (HEART) AND THE OPERATING TABLE CAUSING A CURRENT FLOW OF 50μA FOR A 1000Ω IMPEDANCE -- ENOUGH CURRENT FOR VENTRICULAR FIBRILLATION.

Fig. 17-7. Subject injury caused by a "remote" electrical fault.

response
time of
protection
devices

A catastrophic remote electrical fault may produce a fault current of perhaps 50 amperes which will cause the primary circuit breaker to trip. This circuit breaker may, however, take several hundred milliseconds to disconnect the service after the fault has occurred, thus the fault current may flow for more than 100 milliseconds. As discussed in Section 17.4, if this fault current occurred during the "vulnerable period" of a subject's cardiac cycle, any potential produced between different ground points attached to the subject may be sufficient to cause death.

17.7 OPERATING ROOM ISOLATION

static
charge
prevention

In an effort to reduce the risk of flammable anesthetic ignition in an operating theater, conductive and grounded flooring is used, the subject is connected to a grounded operating table by a conductive rubber mat and moist bedsheet and the operating personnel wear conductive clothing. These precautions prevent the build-up of static electricity and the subsequent spark caused by the discharge of this static electricity.

fault
current
monitors

Due to the presence of this extensive grounding in the operating room, the electrical service to operating rooms normally has both conductors isolated from ground so if a fault develops between one conductor and ground, no fault current can flow. The integrity of this isolation is monitored continuously by ground fault detectors installed between either side of the service and ground. These detectors normally signal any ground fault causing a current to flow in excess of about 1 mA. Unfortunately, there is no more maligned item of electrical equipment in the operating room than the ground fault monitor; if the monitor alarm sounds, the operating room personnel often as not suspect the monitor rather than heed its warning. It should also be noted that the 1 mA level at which the ground-fault monitor alarm sounds may be low enough to protect the subject against currents flowing from surface contacts to his body but is not low enough to protect the subject from shock from internal electrodes or catheters.

17.8 ELECTROCAUTERY AND DEFIBRILLATION

electrocautery

The electrocautery unit is extensively used in operating rooms and is used to provide a source of high-frequency RF current for either cutting tissue or welding tissue together. The electrocautery probe provides a high frequency source of 2 MHz and up to 15 kV with respect to ground. When this probe is applied to a subject during surgery, current flows between this probe and ground. The electrocautery unit is always used in conjunction with a large buttock plate underneath the subject in such a way as to provide a large surface area, good electrical contact, ground to the subject. An electrocautery current of several amperes will thus flow between the electrocautery probe and the buttock plate, however the energy dissipated will cause appreciable heating and burning only in the region of the probe as the current density flowing in the region of the large surface area buttock plate will be quite low.

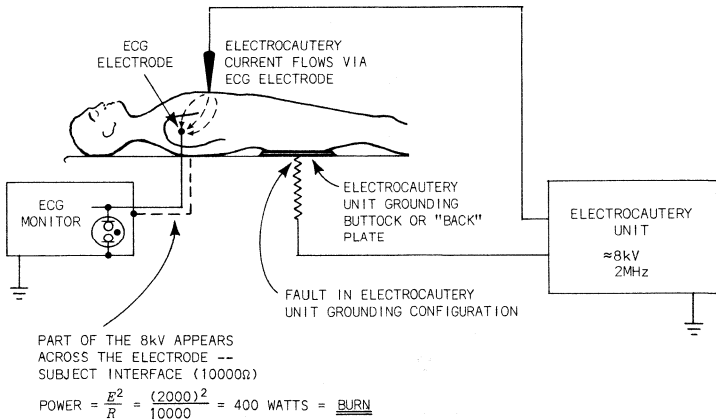


Fig. 17-8. Electrocautery burns at ECG electrode sites.

electrode
burns during
electro-
cautery

If the electrocautery unit does not provide an excellent ground to the subject, either due to poor application or due to a fault within the buttock-plate grounding circuit, then the electrocautery current will be diverted to other grounded points on the subject. It is not uncommon for ECG electrodes to provide this alternative grounding point as all ECG monitoring equipment essentially provides a low impedance path to this high-voltage, high-frequency current. Unfortunately, however, the ECG electrode does not behave in the same manner as the buttock plate as the contact area involved is smaller by a factor of at least 100, perhaps 1,000. The current density is thus increased by a corresponding factor and will almost certainly cause heating and burns at ECG electrode sites. The high-frequency current used in electrocautery is at too high a frequency to cause ventricular fibrillation. In practice, if a surgeon suspects that his electrocautery unit "is not cutting too well," he should be suspicious of the electrocautery unit ground circuitry and should have the electrocautery unit adequately checked before continuing to use it.

defibril-
lation

The defibrillator is another high voltage device commonly used in patient-care areas. The defibrillator provides a single short-duration pulse of up to 10 kV between two large electrodes. These electrodes are normally applied to a subject's chest in cases where the heart is in ventricular fibrillation or has stopped completely. The high energy defibrillation pulse is transmitted between the electrodes and a great portion of this energy is dissipated in the heart. Defibrillation hopefully either re-starts the heart or reverts it from ventricular fibrillation to a normal beating action. Defibrillators should have both electrodes isolated from ground when the high voltage pulse is applied and can normally be expected to provide safe operation if other personnel are not in contact with the subject during the defibrillation process.

Although a defibrillator's output is isolated, this output will be unbalanced capacitively to ground. Thus, unless monitoring devices, such as an ECG instrument, incorporate input circuit protection they may be damaged by the defibrillator pulse.

cardio-
version

If a subject's heart has not stopped beating altogether but is beating erratically, a defibrillator may be operated in a synchronized mode to revert the heart to a normal rhythm. This is referred to as cardioversion. The defibrillator synchronizing pulse is obtained from the subject's ECG R-wave, the defibrillation pulse occurring sometime after the ECG R-wave but avoiding the vulnerable period during the upswing of the ECG T-wave.

TRANSDUCERS - TRANSDUCER SYSTEMS

defined

A transducer may be generally defined as a device which converts one form of energy into another. As far as this text is concerned, a transducer is defined as a device which converts any form of energy other than electrical energy into electrical energy. Transducers are used to measure physical quantities and phenomena by producing an electrical output proportional to this quantity or phenomena. Some transducers require electrical excitation, others produce an electrical output without the aid of excitation. A complete transducer system consists of the transducer as well as the necessary excitation voltage source if excitation is required. Transducer systems may also incorporate some output signal amplification.

biophysical
measurements
only

This chapter should not be regarded as a general reference on transducers, it specifically discusses transducers from a biophysical measurement standpoint and makes no attempt to review some of the more sophisticated transducer systems used for high-accuracy physical measurements. A companion volume entitled *Transducer Measurement Concepts* provides a more general purpose reference source for transducers and transducer measurements.

18.1 RESISTIVE TRANSDUCER CONCEPTS

A resistive transducer is a device whose resistance changes in proportion to some physical quantity. This resistance change is then converted to an electrical output signal by utilizing additional circuitry. Various types of resistive transducers are discussed later in this chapter. For the purpose of this text, resistive transducers may be divided into two broad classifications: Those where the change in resistance expected with a changing physical quantity will be greater than five percent of the transducer's nominal resistance and those where the change in resistance will be less than five percent. This ratio of change in resistance to nominal resistance is normally shown as $\frac{\Delta R}{R}$. This classification is necessary as the circuits used in conjunction with each of these transducer types are quite different.

two
classes

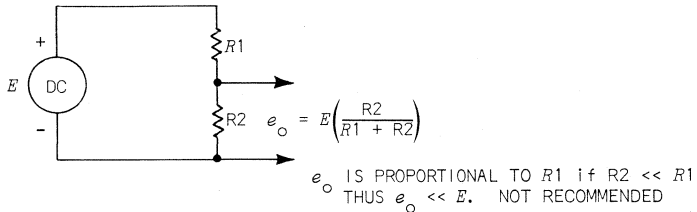
Any one particular transducer may encompass both of the above broad classifications. For example, a thermistor used to measure body temperature must be particularly sensitive and is only expected to be used over a narrow temperature range. This temperature range would not normally change the thermistor's resistance by more than five percent. A thermistor used, however, to record outside air temperature would be expected to operate over a broad temperature range and the thermistor's resistance may be expected to change by 50 percent or more. Thus, the transducer type as well as the transducer's application determines the classification.

type and
application

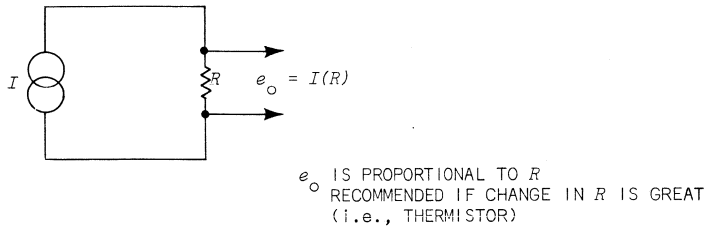
Resistive transducers are normally used in one of the three circuit configurations shown in Fig. 18-1. The circuit shown in Fig. 18-1A essentially powers the resistive transducer from a constant voltage source and monitors the current through the transducer. This circuit is generally the easiest to implement as the excitation source may simply be a low-voltage power supply or battery; however, the circuit is nonlinear unless the value of the current sensing resistor R2 is negligible in proportion to the value of the transducer's resistance R1. The

voltage
excitation

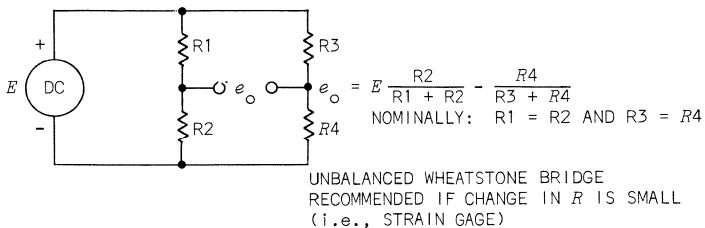
output voltage e_o is thus very small with respect to the excitation voltage E . Since this circuit is somewhat insensitive it can normally only be used in applications where the change in resistance expected with a change in physical quantity will be greater than five percent of the transducer's nominal resistance, i.e. $\frac{\Delta R}{R} > 0.05$.



(A) CONSTANT VOLTAGE



(B) CONSTANT CURRENT



(C) CONSTANT VOLTAGE BRIDGE

NOTE: R = TRANSDUCER
 R = FIXED RESISTOR

Fig. 18-1. Resistive transducer circuits.

current
excitation

The constant current excitation circuit shown in Fig. 18-1B should also be used in applications where the change in resistance expected with a changing physical quantity is greater than five percent of the transducer's nominal resistance. It does not, however, suffer the nonlinearity due to the presence of a current sampling resistor as does the previous circuit. This circuit requires a high-input impedance-recording circuit but this is not normally an important factor with modern electronics. It derives its excitation from a constant current source which may not be as readily available as a constant voltage source.

voltage
excited
Wheatstone
Bridge

The Wheatstone Bridge powered from a constant voltage source shown in Fig. 18-1C is recommended in applications where the change in resistance expected with a changing physical quantity will be less than five percent of the transducer's nominal resistance, i.e. $\frac{\Delta R}{R} < 0.05$. This circuit has the advantage over the previous two circuits discussed in that the output voltage is zero when the transducer's resistance is at its nominal value. The bridge may be, however, somewhat nonlinear as discussed in the following section, thus it should be limited to applications where ΔR is small compared to R .

The feature of providing zero output when the transducer's resistance is at its nominal value is perhaps the Wheatstone Bridge's most desirable characteristic. "Standing" output in the circuits shown in Fig. 18-1A and 18-1B must often be "bucked out" by an offset potential within the amplifier or recording device for the circuits to be useful.

gage
factor

The gage factor of a resistive transducer is the ratio of the relative change in resistance of the transducer to the relative change in the physical quantity being measured. In the particular case of a resistive displacement transducer which, in its simplest form, may consist of a length of resistance wire, displacement can be coupled to the wire to change its length and the gage factor G is computed from the following formula:

$$G = \frac{\frac{\Delta R}{R}}{\frac{\Delta L}{L}} = \frac{\Delta R}{R} \frac{L}{\Delta L}$$

Δl and l refers to the change in length of the resistance wire and the original length of this wire respectively. The term $\frac{\Delta l}{l}$ is referred to as strain, symbol ϵ .

determining
strain and
gage factor

As an example of the above, consider a 1 meter length of resistance wire having a resistance of 100 ohms. If a displacement is mechanically coupled to this wire, the length of the wire may be stretched by one cm to 101 cm. The strain would then be equal to $\frac{1}{100}$ or 0.01. In physical measurement, this strain would be considered large and would be referred to as 10,000 μ strain or 10,000 $\mu\epsilon$. If in stretching the wire, its resistance changes from 100 ohms to, say 102 ohms, then the ratio $\frac{\Delta R}{R}$ is equal to 0.02 and the gage factor G is equal to 2, being the ratio of the resistance change, 0.02, to the strain, 0.01.

18.2 THE UNBALANCED WHEATSTONE BRIDGE

balanced --
unbalanced
bridge

The Wheatstone Bridge circuit was discussed briefly in the previous section, however further discussion of the characteristics of this circuit is desirable due to its extensive use in transducer measurement systems. The Wheatstone Bridge can be used in either of two distinct modes: as a *balanced* bridge where an unknown resistance is measured by adjusting the value of one of the other resistors in the bridge for zero output, or as an *unbalanced* bridge where an unknown resistance is measured by measuring the output voltage produced by bridge imbalance. The balanced bridge is only used in static measurement situations, however the unbalanced bridge is used in both static and dynamic measurement situations. Most transducer applications use the Wheatstone Bridge in its unbalanced condition.

The following paragraph discusses the output voltage obtained from an unbalanced Wheatstone Bridge when one of the resistors comprising the bridge changes its value by ΔR . If only one resistor changes in value, the bridge is said to contain only one active element. This change in value would normally be caused by a change in a physical quantity acting on a transducer.

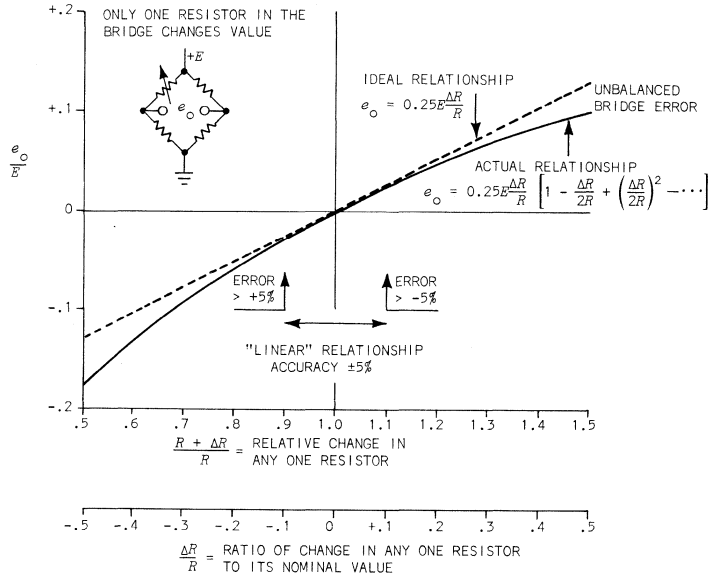


Fig. 18-2. Output voltage from unbalanced Wheatstone Bridge.

calculating
 e_o

The Wheatstone Bridge circuit is shown in Fig. 18-1 together with a formula for determining the output voltage from the circuit. If R_1 and R_2 are equal, and R_3 and R_4 are equal, e_o will be zero. If the resistance of one of the resistors, say R_4 , is changed in value by ΔR , the output voltage may be computed from the equation given in Fig. 18-1. The algebraic solution of this equation reverts to a complex binomial function; however in most instances, second-order terms can be neglected and the equation for the output voltage can be simplified to:

$$e_o = 0.25 E \frac{\Delta R}{R}. \text{ The percentage error is approximately}$$

±5% for ±10% changes in R , i.e., the unbalanced Wheatstone Bridge with one active element is accurate to ±5% for values of $\frac{\Delta R}{R}$ from -0.1 to +0.1. Fig. 18-2

shows both the actual output voltage obtained from an unbalanced Wheatstone Bridge with one active element and the "ideal" output voltage computed from the above equation neglecting second-order binomial terms. The graph presented in Fig. 18-2 confirms the accuracy statement above and also shows that for large ratios of $\frac{\Delta R}{R}$ the unbalanced bridge with one

active element is substantially in error if the second-order binomial terms are neglected. It should be noted that, if the changing resistive element is a thermistor, the curvature of its characteristic largely corrects the nonlinearity of the bridge over the temperature range of clinical interest.

more than
one active
element

Many transducer systems are arranged so that a changing physical quantity causes a resistance change in two of the resistors in the unbalanced Wheatstone Bridge circuit or possibly all four of the resistors in the unbalanced bridge circuit. These bridges are then referred to as having two active elements and four active elements, respectively. Re-analysis of the unbalanced Wheatstone Bridge equations will show that, if the physical quantity to be measured has an opposite effect on both active elements in a bridge containing two active elements, then the output is twice the output of a bridge containing only one active element. Similarly, the output for a bridge containing four active elements is four times the output of a bridge containing only one active element. Re-analysis for two or four active elements will also show that the solution of the equation given in Fig. 18-1 is simple and no second order binomial terms are derived; thus no errors are produced by neglecting such terms. The output voltage formulas for the unbalanced Wheatstone Bridge circuit are summarized below:

If the bridge contains one active element:

$$e_o = 0.25 E \frac{\Delta R}{R} \text{ accurate to } \pm 5\% \text{ for}$$

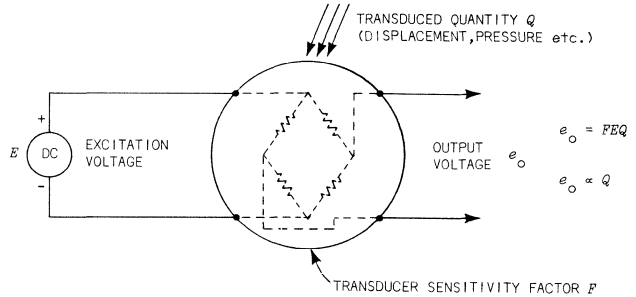
$$-0.1 < \frac{\Delta R}{R} < 0.1$$

If the bridge contains two equally active elements:

$$e_o = 0.5 E \frac{\Delta R}{R} \text{ exactly.}$$

If the bridge contains four equally active elements:

$$e_o = E \frac{\Delta R}{R} \text{ exactly.}$$



EXAMPLE: PRESSURE TRANSDUCER WITH SENSITIVITY FACTOR SUPPLIED WITH TRANSDUCER OF $118\mu\text{V}/\text{V}/\text{cmHg}$ PRESSURE.
 SAY, CONVENIENT EXCITATION VOLTAGE OF 6V DC
 e_o FOR ONE cmHg PRESSURE = $FEQ = 118 \times 10^{-6} \times 6 \times 1 = .708\text{mV}$
 THUS, WITH 6.0V OF EXCITATION, SYSTEM GIVES
 $.708\text{mV}$ OF e_o/cmHg PRESSURE.

ALTERNATIVELY: ADJUST EXCITATION VOLTAGE FOR A MORE CONVENIENT CALIBRATION FACTOR.

SAY WE WISH 1mV OF e_o/cmHg PRESSURE;

THEN, $e_o = FEQ$

$$1 \times 10^{-3} = 118 \times 10^{-6} \times E \times 1$$

$$E = 8.49\text{V}$$

THUS, WITH 8.49V OF EXCITATION, THE SYSTEM GIVES
 1.0mV OF e_o/cmHg PRESSURE.

THE EXCITATION VOLTAGE MUST BE WITHIN THE TRANSDUCER'S RATING. FOR MOST TRANSDUCERS THE EXCITATION RANGE IS 5V TO 8V .

Fig. 18-3. Transducer system calibration using sensitivity factor F supplied with transducer.

18.3 PRACTICAL TRANSDUCER SYSTEMS USING THE UNBALANCED WHEATSTONE BRIDGE

sensitivity factor F

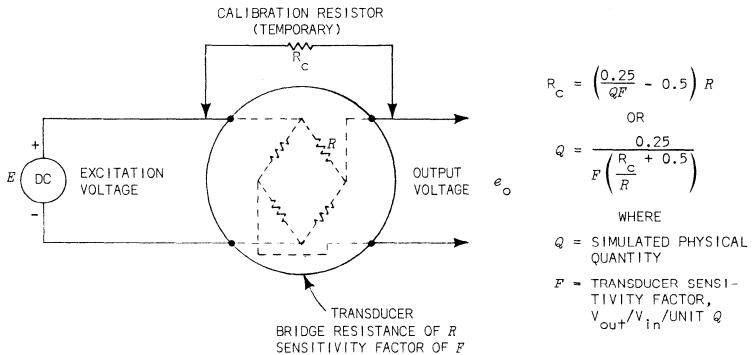
Practical transducers are self-contained units that produce an electrical output proportional to the excitation voltage and the physical quantity to be measured. Although the transducer may internally consist of a Wheatstone Bridge having one or more active elements, it is unnecessary to know the characteristics of the elements comprising this bridge if a calibration factor or transducer sensitivity factor F is supplied with the transducer at the time of purchase. The transducer sensitivity factor F is normally given as volts output per volt of excitation per unit of physical quantity to be measured. The actual output voltage is determined as shown in Fig. 18-3.

If the transducer sensitivity factor for a particular transducer is unknown, it may be determined by calculating
 subjecting the transducer to a known amount of the

known source, and the output voltage measured accurately with a calibrated oscilloscope or voltmeter, the sensitivity factor may be calculated by rearranging the equation shown in Fig. 18-3.

calibration
resistor

With the above system, an accurate measurement of the physical quantity under investigation depends on the accuracy of the transducer sensitivity factor, the accuracy of the excitation voltage, and the accuracy of the output voltage indication device. Many transducer systems incorporate a calibration resistor as shown in Fig. 18-4. This calibration resistor is chosen to have the same effect on the transducer as would a certain amount of the physical quantity to be measured. It is thus only necessary to connect this calibration resistor as shown in Fig. 18-4 and to adjust the output voltage indication device for a known deflection. This resistor fully "calibrates" the transducer system and it is unnecessary to know the exact value of the excitation voltage or the exact sensitivity of the output voltage indication device. Many transducers are supplied by manufacturers at the time of purchase complete with a calibration resistor. If the transducer sensitivity factor F is known, an appropriate calibration resistor value can be selected as shown in Fig. 18-4.



BRIDGE RESISTANCE R IS THE NOMINAL RESISTANCE OF EACH OF THE FOUR ARMS OF THE BRIDGE. IT MAY BE OBTAINED FROM THE MANUFACTURER'S DATA SHEET FOR THE TRANSDUCER OR MEASURED WITH AN ACCURATE RESISTANCE BRIDGE BY SIMPLY MEASURING THE RESISTANCE BETWEEN THE EXCITATION TERMINALS WITH THE OUTPUT TERMINALS OPEN CIRCUITED OR VICE VERSA.

EXAMPLE: PRESSURE TRANSDUCER SENSITIVITY FACTOR SUPPLIED WITH TRANSDUCER OF $118\mu V/V/cmHg$ PRESSURE -- BRIDGE RESISTANCE 350 Ω

WE WISH TO SIMULATE A PRESSURE OF 10cmHg:

$$R_c = \left(\frac{0.25}{QF} - 0.5 \right) R = \left(\frac{0.25}{10 \times 118 \times 10^{-6}} - 0.5 \right) 350 = 74.2k\Omega$$

Fig. 18-4. Simulation of transduced quantity with a calibration resistor.

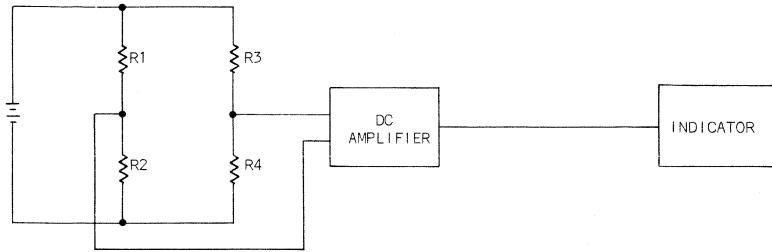
18.4 AC- AND DC-BRIDGE SYSTEMS

DC-bridge
systems

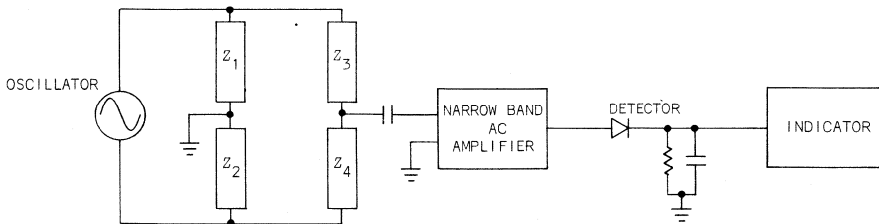
The previous discussion applies to transducer systems having either DC or AC excitation sources. In practice, both DC-bridge and AC-bridge systems are used. The DC-bridge systems shown in Fig. 18-5 simply consist of a battery or DC power supply as the excitation source and incorporate a DC-coupled amplifier in the output voltage indication system. Since the excitation is DC, this system cannot be used with reactive transducers; that is, transducers where the active elements consist of inductors or capacitors rather than resistors. Until recently this DC-bridge system was not preferred for high-accuracy or high-sensitivity measurements due to the relative instability of DC-coupled amplifiers. Stable DC-coupled amplifiers, such as the Tektronix Type 3A9 Differential Amplifier, are now available and thus DC-excited bridge systems are regaining usage.

simple AC-
bridge
systems

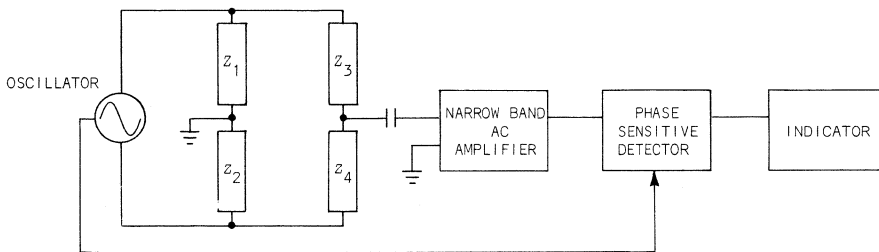
Perhaps the most economical bridge system is the simple AC-bridge system shown in Fig. 18-5. With this system, the bridge is excited by a sinewave oscillator, normally within the audio-frequency range, and the output is amplified via either a wide-band AC-coupled amplifier or, preferably, a narrow-band AC-coupled amplifier. Since the amplifier is AC coupled, DC stability is unimportant. The amplified bridge output signal is detected with a simple detector, however, as this detector is not phase sensitive, this simple AC-bridge system will provide essentially "positive" output for both "positive" and "negative" changes away from the balance condition in the physical quantity to be measured. Often, however, the nature of this physical quantity overcomes this apparent ambiguity.



(A) DC SYSTEM



(B) SIMPLE AC SYSTEM



(C) PHASE SENSITIVE AC SYSTEM

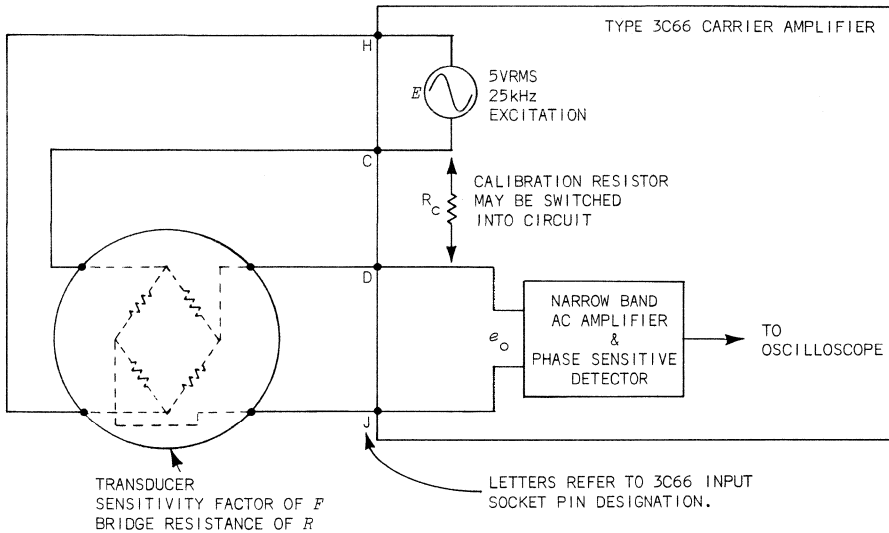
Fig. 18-5. DC and AC bridge systems.

carrier
amplifier
systems

The detector used in the simple AC system discussed in the previous paragraph may be replaced with a phase-sensitive detector that solves the apparent ambiguity by producing both positive and negative outputs. Such a system is referred to as a phase-sensitive AC system or a carrier amplifier system as shown in Fig. 18-5. The Tektronix Type 3C66 plug-in unit for the Type 561B Oscilloscope or 564B Storage Oscilloscope employs this phase-sensitive AC-bridge system.

3C66

The Tektronix Type 3C66 Carrier Amplifier was primarily intended for use with strain gages and is thus calibrated in microstrain or $\mu\epsilon$. This instrument does, however, incorporate switching circuitry to allow a built-in $150\text{-k}\Omega$ shunt calibration resistor to be used. Thus the μ STRAIN/DIV switch may be regarded as a variable gain control on the transducer's output voltage detection circuitry and a calibration resistor may be used to calibrate the system as discussed in Section 18.3. The Tektronix Type 3C66 Carrier Amplifier system is shown in Fig. 18-6. The built-in $150\text{-k}\Omega$ shunt calibration resistor in the Type 3C66 may be replaced with any desired value of calibration impedance. The calibration impedance must be of the same form as the impedance that is changing in the transducer, thus resistive transducers require calibration resistors, inductive transducers require calibration inductors and capacitive transducers require calibration capacitors.



SELECT A CALIBRATION RESISTOR AS DESCRIBED IN FIG. 18-4. PLACE THIS CALIBRATION RESISTOR INSIDE THE TYPE 3C66 ON THE TERMINALS AS DESCRIBED IN THE TYPE 3C66 MANUAL. ADJUST TYPE 3C66 μ STRAIN/DIV SWITCH AND GAIN CONTROL FOR A CONVENIENT DEFLECTION.

- EXAMPLE: PRESSURE TRANSDUCER - SENSITIVITY FACTOR SUPPLIED WITH TRANSDUCER OF $118\mu\text{V}/\text{V}/\text{cmHg}$ PRESSURE - BRIDGE RESISTANCE OF 350Ω
- SELECT A CALIBRATION RESISTOR TO SIMULATE A PRESSURE OF, SAY, 10cmHg . FROM FIG. 18-4, $R_C = 74.2\text{k}\Omega$.
 - INSTALL THIS $74.2\text{k}\Omega$ RESISTOR INTO THE 3C66.
 - ADJUST TYPE 3C66 CONTROLS FOR CONVENIENT DEFLECTION, SAY, 2 DIVISIONS, WITH μ STRAIN/DIV AT 500.

NOTE CALIBRATION: $10\text{cmHg} = 2 \text{ DIVISIONS AT } 500\mu\text{STRAIN/DIV}$
 $\therefore 10\text{cmHg} = 1000\mu\text{STRAIN}$
 $\therefore \text{SENSITIVITY} = 100\mu\text{STRAIN}/\text{cmHg PRESSURE}$

Fig. 18-6. Tektronix Type 3C66 Carrier Amplifier.

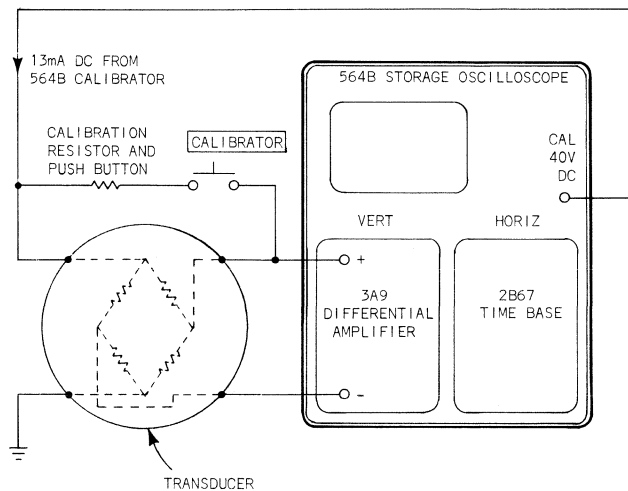


Fig. 18-7. Transducer system using conventional oscilloscope and plug-in units.

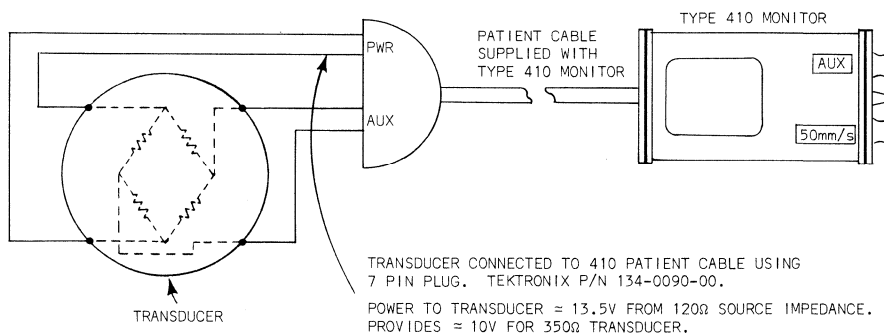


Fig. 18-8. AC coupled transducer system with Type 410 Monitor.

564B
system

A DC-transducer system may be fabricated using a Tektronix Type 564B Storage Oscilloscope with a 3A9 Vertical Amplifier and 2B67 Time Base unit as shown in Fig. 18-7. This system excites the transducer from an approximate 13 mA current source, however since the transducer's impedance is essentially constant, the voltage between the transducer's excitation terminals will be constant and the system will operate as if it were excited from a constant voltage source. This system is suitable for use with transducers having a nominal bridge resistance of between 100 ohms and 2,000 ohms.

410
system

The transducer system shown in Fig. 18-8 may be used in some limited applications where the physical quantity to be measured is essentially an AC signal. This system is thus suitable for recording the difference between diastolic and systolic pressures using a pressure transducer, however since the Type 410 is insensitive to DC signals, it will not show the absolute values of these pressures.

18.5 DISPLACEMENT TRANSDUCERS

A displacement transducer converts linear or rotational displacement into an electrical output. Various forms of resistive displacement transducers are shown in Fig. 18-9. Each of these transducers changes resistance due to displacement; this resistance change is then converted to an electrical output by using any one of the circuits shown in Fig. 18-1.

potenti-
ometer

Perhaps the simplest resistive displacement transducer is the common potentiometer. The resistance of the potentiometer changes with rotational displacement which may be obtained from a linear displacement via either a lever system or a pulley. This type of resistive displacement transducer is commonly used in spirometers as discussed in Section 9.4.

fluid-filled rubber tube The fluid-filled rubber tube shown in Fig. 18-9 changes its cross-sectional area and length due to a linear displacement or force acting on the rubber tube. This changing cross-sectional area and length causes a change in the resistance of a conductive fluid within the tube as measured between each end of the tube. Mercury has historically been used as the conductive fluid in this type of transducer, however the resistance of mercury is extremely low and special techniques are required to detect any change in this resistance. Fluids with much higher resistivities are preferred and are indeed used in many commercial fluid-filled rubber-tube transducers.

bonded strain gage The three strain-gage resistive-displacement transducers shown in Fig. 18-9 represent the basis for most modern transducers. The bonded strain gage consists of a length of fine resistance wire bonded to a backing of paper or epoxy. As linear displacement is coupled to this backing material, it stretches and the resistance of the wire bonded to this material therefore increases. The unbonded strain gage is somewhat similar to the bonded strain gage with the exception that the resistance wire is self supporting as shown in Fig. 18-9. The bonded semiconductor strain gage is also somewhat similar to the bonded strain gage with the exception that the resistance wire element is replaced with a filament of semiconductor material, usually silicon.

unbonded strain gage

semiconductor strain gage The principal advantage of the bonded semiconductor strain gage is its high gage factor, in the order of 120 compared to 2 or perhaps 3 for the bonded or unbonded resistance-wire strain gages.

displacement inertia With any of the displacement transducers shown in Fig. 18-9, the mechanical source providing the displacement must also provide some force to overcome inertia in the transducer. The fluid-filled tube may require considerable force in conjunction with the linear displacement to be able to stretch the rubber tube. The unbonded strain gage requires some force to stretch the resistance wires comprising the gage. Such devices should more correctly be classed as force transducers that may be used for displacement measurements if the force driving the system is large compared with that required to drive the transducer.

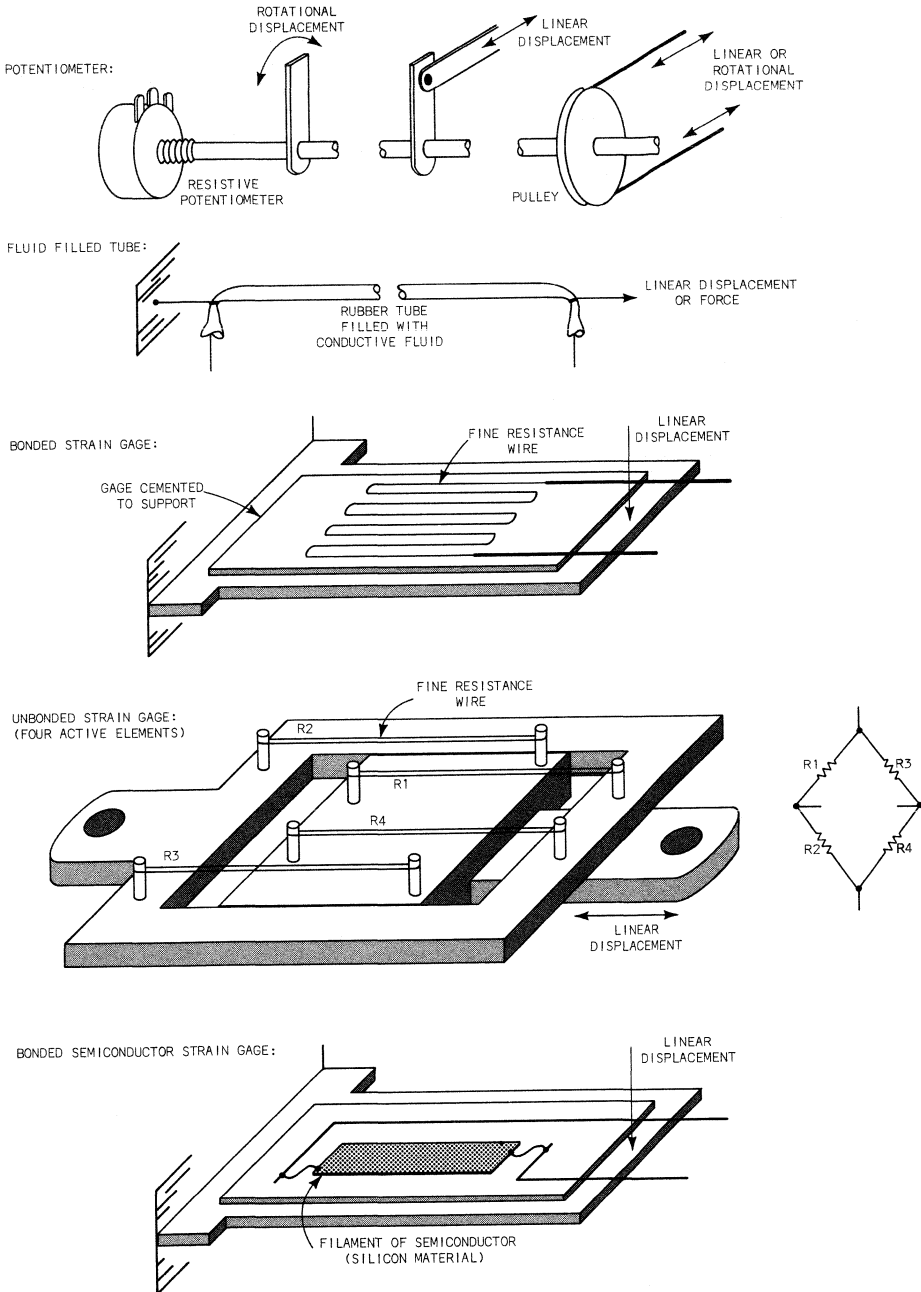


Fig. 18-9. Resistive displacement transducers.

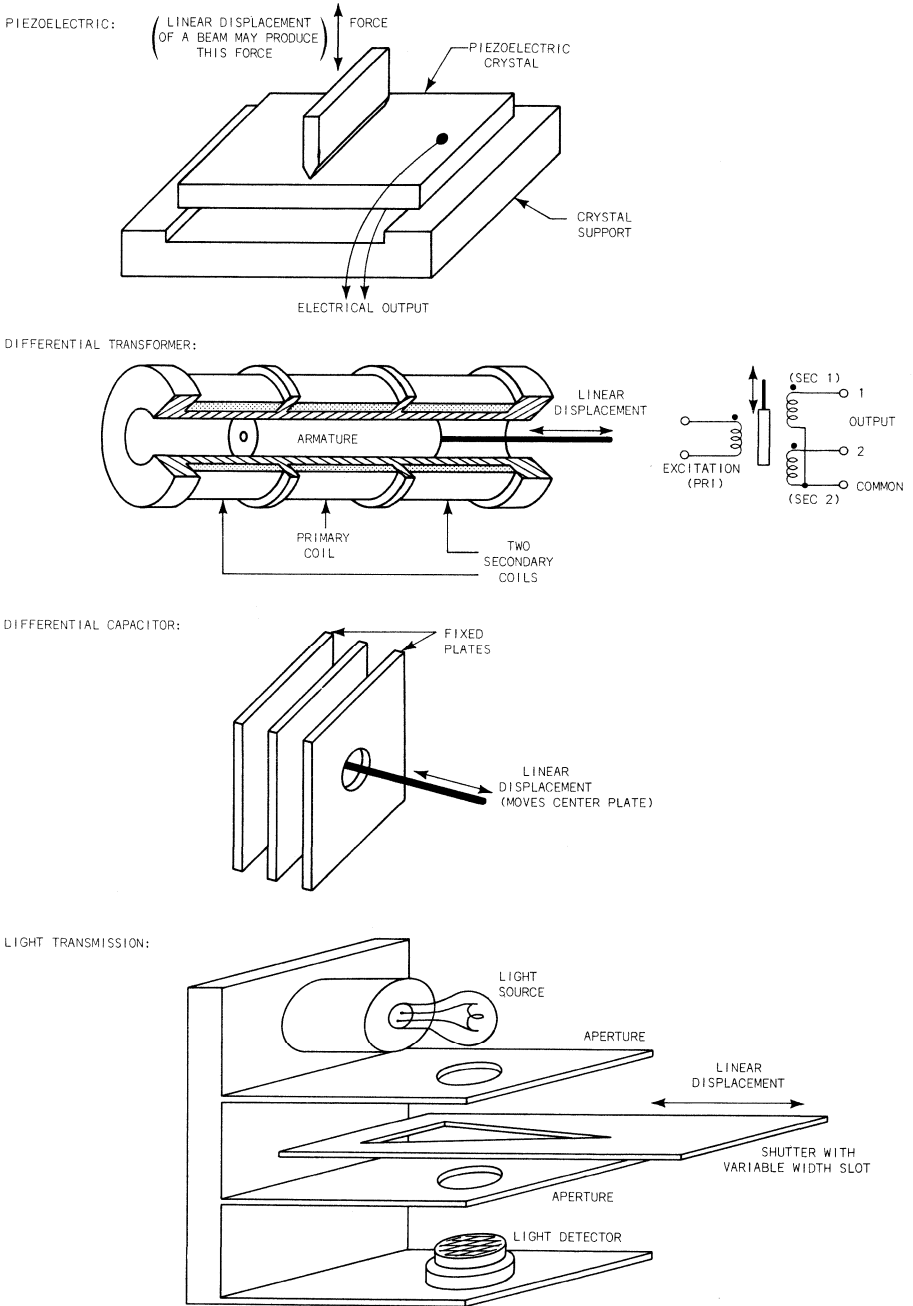


Fig. 18-10. Nonresistive displacement transducers.

piezo-
electric

Various forms of nonresistive displacement transducers are shown in Fig. 18-10. The piezoelectric force transducer produces an electric charge between the faces of a solid crystal when this crystal is subjected to bending which may be caused by displacement of a beam, etc. A principal advantage of the piezoelectric crystal transducer is that no excitation source is required, however displacement does produce a charge rather than a voltage and this charge is normally quickly dissipated by the input impedance of the amplifier used in conjunction with this transducer. It is thus primarily suited to the measurement of AC quantities, however the transducer can be used in conjunction with a charge amplifier to provide good low-frequency response characteristics. A piezoelectric displacement transducer is used in the Korotkoff sound microphone shown in Chapter 8, Fig. 8-10.

differential
transformer

A differential-transformer displacement transducer is shown in Fig. 18-10. The differential transformer consists of three inductors coupled magnetically by a common armature. One of these three inductors is excited from an AC source (primary winding) and this excited winding induces voltages into the other two windings (secondary windings), which are connected in opposition. The induced voltage is dependent on the amount of coupling and thus on the relative position of the armature. With the armature coupling energy equally from the primary winding to each of the secondary windings, voltages of equal amplitude appear across each winding. When the core is moved off center, however, each secondary is not linked identically. One secondary produces a larger voltage than the other and thus the potentials produced in the secondaries do not cancel and an output voltage is obtained. The two secondary windings of a differential transformer may be considered as two elements of an unbalanced Wheatstone Bridge with the other two elements consisting of fixed resistors. The differential transformer is particularly suited for use with either of the AC-bridge systems shown in Fig. 18-5. They are also available in DC-to-DC versions with all the necessary electronics built into the actual transducer.

differential capacitor The differential-capacitor displacement transducer shown in Fig. 18-10 essentially consists of two capacitors arranged so that a common plate for both of these capacitors is coupled to a linear displacement. Any movement of this common plate increases the capacitance in one capacitor while simultaneously decreasing the capacitance in the other. These two capacitors may be used to form two elements in an AC Wheatstone Bridge system with the other two elements consisting of fixed resistors.

The light-sensitive displacement transducer shown in Fig. 18-10 employs a variable-width slot coupled to the displacement. Light is passed through this slot, the width of the slot determining the amount of light that can pass. The varying light intensity is normally detected with a light-sensitive transducer as discussed in Section 18.7.

18.6 FORCE, PRESSURE AND ACCELERATION TRANSDUCERS

force Force, pressure and acceleration transducers are related to displacement transducers as shown in Fig. 18-11. A force transducer converts force to linear displacement using a spring or some other resilient material. Force on the transducer produces stress in the sensing element. This stress causes strain and the strain is then detected with a strain gage displacement transducer.

pressure The pressure transducer shown in Fig. 18-11 is basically a force transducer with fluid pressure being converted to force via a diaphragm. Since the force produced is proportional to the fluid pressure, the transducer's resistance change is proportional to the fluid pressure.

acceleration The acceleration transducer shown in Fig. 18-11 simply consists of a force transducer coupled to a known mass. This mass converts an acceleration to a force. Although the acceleration transducer shown in Fig. 18-11 is sensitive to acceleration in one direction only, modifications to this principal allow acceleration transducers sensitive to acceleration in any direction to be constructed. The weight transducer shown in Fig. 18-11 is essentially a force transducer with the force being proportional to the weight of a body applied to the transducer and to the gravity constant g .

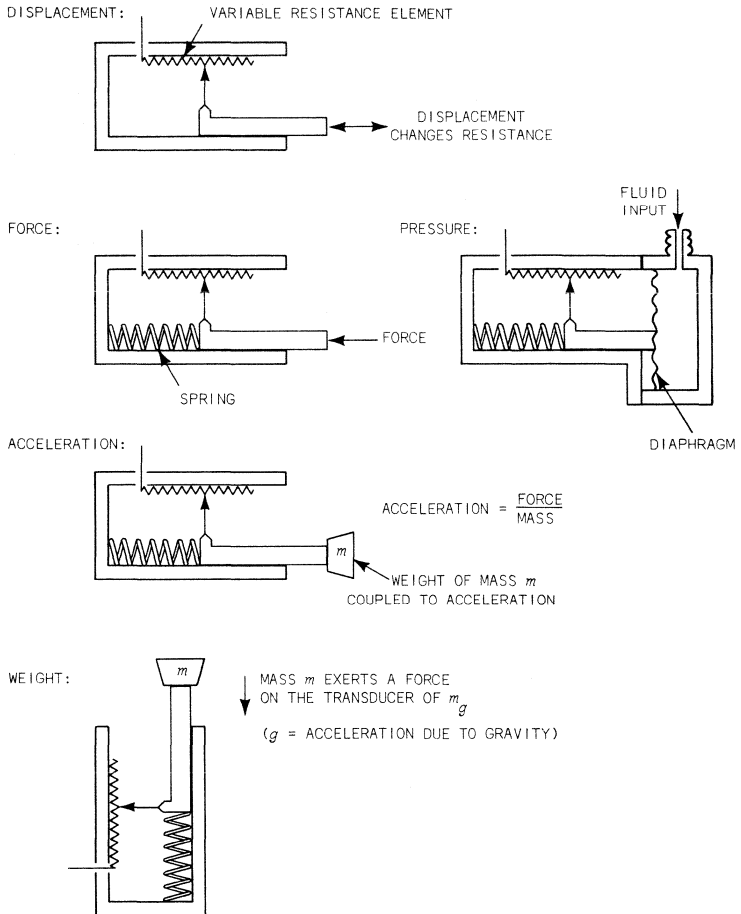


Fig. 18-11. Transducer evolution.

velocity
pickups

Velocity pickups, consisting simply of a coil located in a magnetic field, are commonly used for vibration pickups and in dynamic microphones. Displacement and acceleration may be obtained from the output of a velocity transducer by integrating or differentiating this output voltage. Conversely, displacement can be measured with an acceleration transducer by double integration of the output voltage and acceleration can be measured with a displacement transducer by double differentiation.

The mathematical relations involved are shown here.

Differentiation processing:

$$v = \frac{dx}{dt} \quad x = \text{Displacement}$$

$$a = \frac{dv}{dt} = \frac{d^2x}{dt^2} \quad v = \text{Velocity}$$

$a = \text{Acceleration}$

$t = \text{Time}$

Integration processing:

$C = \text{Constant}$

$$v = C1 \int a dt$$

$$x = C2 \int v dt = C1C2 \int a dt$$

For sinusoidal motion, the above may be simplified to:

$$v_{\text{peak}} = 2\pi f x_{\text{peak}}$$

$$a_{\text{peak}} = (2\pi f)^2 x_{\text{peak}} \quad f = \text{Frequency of sinusoidal motion}$$

Calibration of systems using this indirect method can be difficult, thus these methods are primarily used when uncalibrated or relative measurements are satisfactory. Double differentiation can often result in excessive noise and thus should be avoided.

pressure
transducer
predominant

volume
displacement

Although force, pressure, acceleration and weight transducers are all used in biophysical measurements, the pressure transducer is of particular interest due to its extensive use for monitoring the activity of the circulatory system and respiratory system as discussed in Chapters 8 and 9. An important characteristic of a pressure transducer used for biophysical measurements is its volume displacement, that is, the amount of fluid that must flow into the pressure transducer to produce movement of the diaphragm within the transducer. Volume displacement in pressure transducers is normally measured in cubic millimeters per 100 mm Hg pressure change. Volume displacements range from below .01 cubic mm per 100 mm Hg pressure to above 20 cubic mm/100 mm Hg pressure. Transducers with high volume displacements normally have poor high-frequency response characteristics due to the finite time required to move the large volume of fluid involved. Transducers for blood pressure measurement should have a volume displacement of less than .04 cubic mm/100 mm Hg pressure.

18.7 TRANSDUCERS FOR NONMECHANICAL QUANTITIES

The range of physical quantities that can be measured with the aid of a transducer is almost limitless, as is the range of transducers that may be useful in the biophysical measurement sciences. Transducers are used extensively in the biophysical sciences to measure the following nonmechanical quantities: temperature, light, radiation, liquid chemical composition and gaseous chemical composition. These transducer types are discussed briefly below, however more detailed information is available from the principal transducer manufacturers; perhaps some of the more prominent are Statham Instruments, Inc., Oxnard, California; Beckman Instruments, Inc., Palo Alto, California; E & M Instrument Company, Inc., Houston, Texas and many others. An excellent reference for the application of transducers to biomedical instrumentation is Chapters 1 through 9 of *Principles of Applied Biomedical Instrumentation* by L. A. Geddes and L. E. Baker, published by John Wiley and Sons, Inc., New York, N.Y.

thermistor

Temperature transducers can be divided into two categories; temperature dependent resistors and thermoelectric generators. The thermistor is a temperature dependent resistor whose resistance decreases as temperature is increased. A typical thermistor may be expected to produce a one percent change in resistance for a temperature change of one or two degrees within normal environmental temperature ranges. For the resistance change to be detected, some form of resistance detection circuitry is necessary. It is important that the power level involved in this circuitry does not cause self-heating within the thermistor.

thermocouple

A thermocouple consisting of a junction between two dissimilar metals is a thermoelectric generator, however, this form of temperature transducer is rarely used for biophysical measurements due to its relative insensitivity compared to thermistors.

light
sensitive
resistor

Optical transducers sensitive to light are used in the mechanical displacement transducer shown in Fig. 18-10 and in the plethysmographs shown in Chapter 8. Although many different forms of optical transducers are available, the most common type used in biophysical measurements is the light-sensitive resistor. Light-sensitive resistors, or photo resistors, consist of a thin film of resistive material deposited on the inside of an evacuated glass or plastic chamber. The most commonly used material is cadmium sulphide, however, cadmium selenide and semiconductor materials are used in certain applications. The response of these photo resistors is reasonably linear; the resistance decreases proportionally with an increase in the level of illumination. A typical "dark" resistance would be above 10 megohms, this resistance would decrease to a few kilohms with normal room illumination of about 100 foot-candles.

radiation
detection

Radiation transducers are used in biophysical measurements to check for the possibility of a radiation hazard. Radiation transducers may detect one or more of the three principal forms of radiation: alpha particles, beta particles and gamma rays. Gamma radiation is usually considered the most dangerous because of its higher energy content. Radiation transducers either detect radiation directly or detect light emitted by certain crystals when exposed to radiation. Perhaps the most common form of radiation detector is the Geiger-Muller counter which, when subject to radiation produces ionization in a sealed gas-filled tube which allows current to flow between electrodes placed in the tube.

chemical
detection

Liquid chemical transducers are used in biophysical measurements to provide a continuous chemical analysis of body matter, particularly blood. Liquid chemical transducers are used to measure the pH or relative acidity of body fluids and are also used for routine blood analysis to determine the amount of dissolved oxygen or carbon dioxide in the blood. In addition, liquid chemical transducers are available to determine the relative amounts of potassium and sodium ions in body fluid. Gaseous chemical transducers are used to analyze the gaseous content of expired air, particularly the amounts of oxygen and carbon dioxide, and also are used as oxygen detectors in atmospheric warning systems.

AMPLIFIERS

Physiological signals acquired by either electrodes or transducers are typically below 10 mV in amplitude and must therefore be amplified to be compatible with display devices and recorders. This chapter discusses the characteristics of amplification systems as seen by electrodes. The discussion will be limited to the process of amplification, additional signal processing is discussed in Chapter 20. In practice, most physiological amplifiers include both amplification and signal processing within the one package. For the purpose of this discussion, therefore, an amplifier is defined as a device having a high input impedance and a low output impedance and providing either a fixed or perhaps a variable voltage gain. The amplifier may or may not contain DC-offset capabilities to offset the DC level of the output with respect to the input DC level.

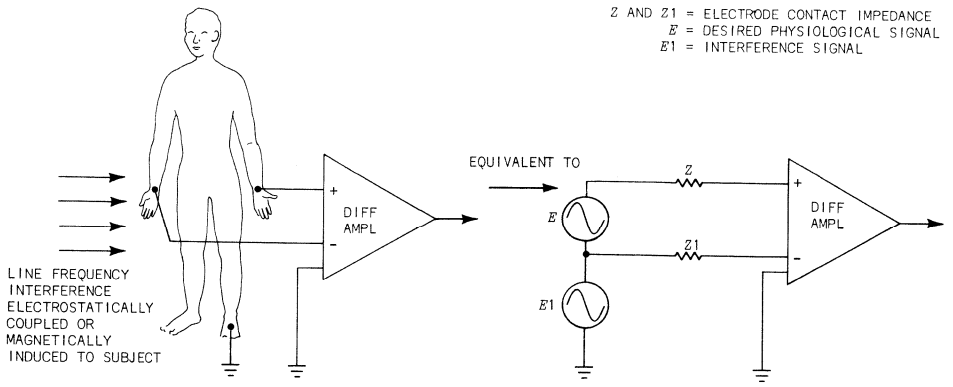


Fig. 19-1. Typical physiological measurement configuration.

19.1 THE DIFFERENTIAL AMPLIFIER USED WITH ELECTRODES

A typical physiological measurement configuration is shown in Fig. 19-1. In almost all physiological measurement situations, two signals are produced by the subject: a desired physiological signal such as the ECG or EEG and an interference signal, typically 60 Hz, due to electrostatically coupled or magnetically induced interference. The purpose of the differential amplifier is to reject this interference signal and to amplify the desired signal. Two impedances, Z and Z_1 , representing electrode contact impedances are shown in Fig. 19-1. The source of these impedances is discussed in Chapter 16 when referring to the characteristics of various electrode systems. Z and Z_1 will normally lie in the range from 1,000 ohms to 50,000 ohms and it is possible for the difference between Z and Z_1 in any one system to be as great as 10,000 ohms. Although all modern differential amplifiers have input impedances above 1 megohm, these finite impedances Z and Z_1 cannot be ignored as will be seen later in this chapter. The desired physiological signal shown in Fig. 19-1 is connected so that it appears between the two input terminals of the differential amplifier and is referred to as the differential signal. The interference signal shown appears between both inputs of the differential amplifier and ground and is referred to as the common mode signal.

19.2 COMMON MODE REJECTION

As stated in the previous paragraph, the purpose of a differential amplifier is to reject common mode signals. The "common mode rejection ratio" or CMRR of an amplifier is its relative ability to reject common mode signals. The common mode signal is defined as the signal that is applied to both inputs in the same phase, that is:

$$\text{common mode signal} = \frac{(\text{signal to } + \text{ input}) + (\text{signal to } - \text{ input})}{2}$$

Common mode rejection ratio is defined as the ratio between the amplitude of the common mode signal to the amplitude of an equivalent differential signal that would produce the same output from the amplifier. As an example, consider a differential amplifier with a single 1-V source connected from ground to both inputs of the amplifier, that is, in common mode. If the output signal is equivalent to that produced by a 10 μV differential signal, the common mode rejection ratio would then be the ratio of this apparent differential signal to the true common mode signal; that is, 1V:10 μV or 100,000:1. Common mode rejection is often specified in decibels; a CMRR of 100,000:1 being referred to as a common mode rejection of -100 dB. The relationship is --

$$\text{Common Mode Rejection in dB} = 20 \log_{10} \text{CMRR}$$

CMRR versus
source
impedance

The common mode rejection ratio calculated previously was obtained by applying the same one volt common mode signal to both inputs of the differential amplifier. Let us now consider the case where an impedance of 150 ohms is inserted between the one volt source and the positive input of the differential amplifier with the negative input of the differential amplifier connected directly to the source as previously. Considering the positive input circuit and assuming an input impedance to ground from this positive input of 10 megohms, the 150-ohm resistor in conjunction with the 10-megohm input resistance of the differential amplifier will form a small but nevertheless highly significant voltage divider. Now, instead of the full one volt of common mode signal reaching the differential amplifier, 15 μV appear across the 150-ohm resistor and only one volt minus 15 μV or .999985 volts reach the positive input of the amplifier. Since no resistor is inserted in series with the negative input, the full one volt common mode signal reaches this negative input. A common mode signal is therefore $\frac{1 + .999985}{2} = .9999925$ volts. The signal at the +input is 7.5 μV less than this and the signal at the -input is 7.5 μV greater than this. The additional 15 μV on the negative input with respect to the positive input is a differential signal and will be amplified as a differential signal. Assuming a gain of 20,000, this 15 μV will appear as 300 mV at the output.

It is thus apparent that the amplifier described in the preceding paragraph with a CMRR of 100,000:1 appears to produce an additional output due to the 150-ohm "source impedance unbalance." If 200 mV of common mode signal appear at the output due to the CMRR of the amplifier and 300 mV of common mode signal appear at the output due to source-impedance unbalance, a maximum of 500 mV of common mode signal may appear at the output. These two signals would normally not add directly, but being AC signals, they must be added algebraically and it is conceivable that they may in fact tend to cancel each other. An equivalent differential signal that would produce this same output would be $\frac{500 \text{ mV}}{20,000}$ or 25 μV . The apparent CMRR is then the ratio of this equivalent differential signal to the true common mode signal; that is .9999925V:25 μV or $\approx 40,000:1$. It is thus apparent that the 150-ohm source-impedance unbalance has reduced the CMRR of the amplifier from 100,000:1 to an apparent CMRR of 40,000:1. As these signals add algebraically and may tend to cancel each other, 40,000:1 will be a worst-case CMRR. In practice the apparent CMRR may be increased by the source impedance unbalance.

Extension of the previous discussion will show that it is not the absolute value of the source impedance that determines this apparent reduction in CMRR but rather the difference between the source impedances (source impedance unbalance) between the positive and negative inputs of the differential amplifier. As stated in the preceding section, it is possible for a source-impedance unbalance of up to 10,000 ohms to occur when recording physiological signals. It is thus desirable either to attempt to minimize this unbalance, or to increase the input impedance of the differential amplifier to the common mode signal in an effort to increase the ratio between the source impedance unbalance and the common mode input impedance of the differential amplifier.

From the preceding discussion it is apparent that the output of any differential amplifier will consist of three components --

1. A desired output component due to amplification of a differential signal by the differential amplifier,

source
impedance
reduces
CMRR

source
impedance
unbalance

output has
three
major
components

2. An undesired component due to incomplete rejection of common mode interference signals by the differential amplifier and,
3. An undesired component due to source impedance unbalance allowing a small proportion of a common mode signal to appear as a differential signal to the amplifier.

These three output components are shown in Fig. 19-2 together with the equations used for determining the values of each of these three output-signal components.

fourth
component

When a differential amplifier is used on a subject there will also be a fourth output; an undesired output component due to amplification of an *undesired* differential input from the subject, due to the subject's finite resistance between the input terminals and the presence of AC interfering currents through this resistance. This fourth output limits the useful rejection ratio of an ECG amplifier to about 100,000:1.

calculation
of output
components

Referring back to the amplification system discussed earlier in this section, this system is illustrated in Fig. 19-3 with the one volt common mode signal consisting of a one volt P-P, 60-Hz sinewave and a desired differential signal consisting of a 100 μ V P-P, 2-Hz sinewave. Applying the formula shown in Fig. 19-2 to this amplification system gives a desired output of 2 volts of 2-Hz sinewave and an undesired output of 0.5 volts of 60-Hz sinewave. This undesired output consists of 0.2 volts due to the inherent CMRR of the amplifier and 0.3 volts due to the 150-ohm source-impedance unbalance of the system.

When measuring the CMRR of modern amplifiers, it is necessary to use about one volt of common mode signal so that the output due to this signal can be discerned from noise. (1 V common mode with 100,000:1 CMRR is equivalent to a 10 μ V differential input signal.) Typical common mode signals in biophysical measurements are from 1 mV to 100 mV. The CMRR to these signals will be slightly greater than the CMRR to a one volt signal due to small nonlinearities in the amplifier's input devices.

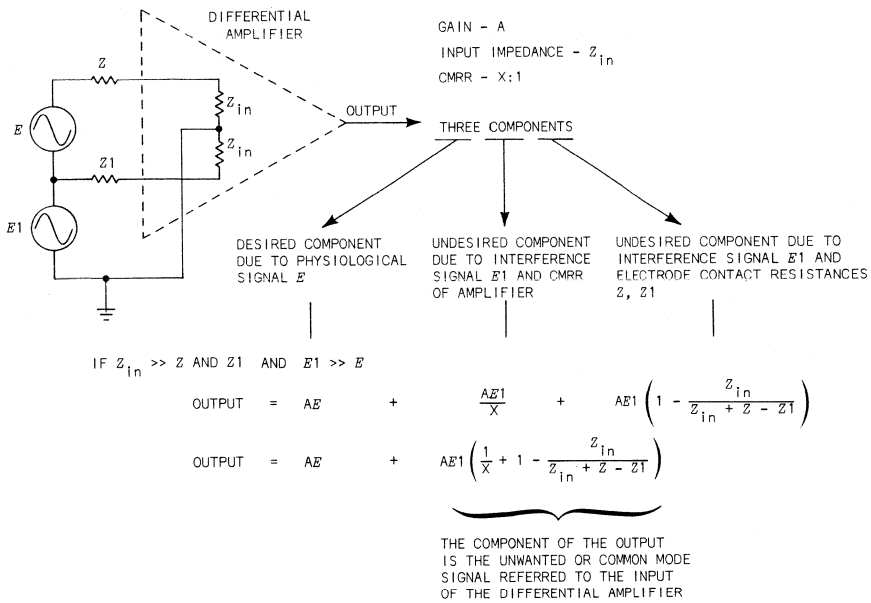


Fig. 19-2. Common mode rejection.

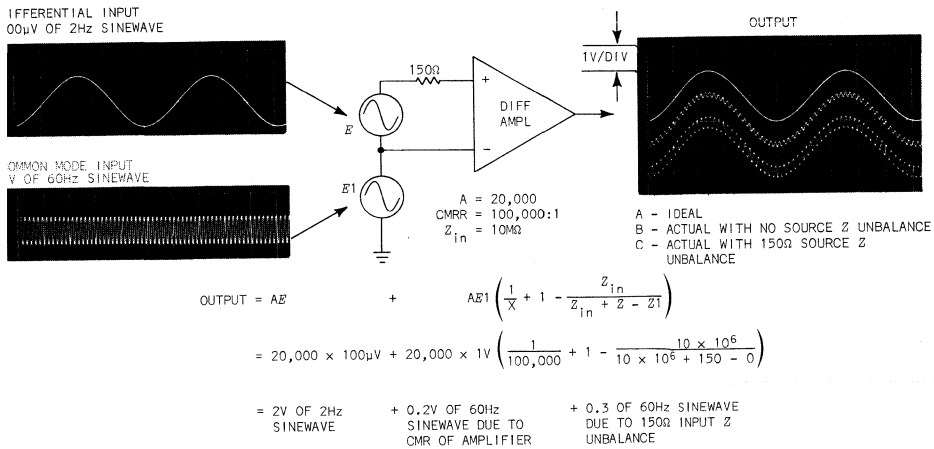


Fig. 19-3. Example of common mode rejection calculation.

coupling
capacitor
unbalance

The principal common mode signal encountered in biophysical recording is line frequency interference at either 50 Hz or 60 Hz. Let us now examine the validity of inserting coupling capacitors in series with the differential amplifier inputs to provide an AC-coupled amplifier. In physiological situations these capacitors may be intended to block the DC-offset potentials encountered in the electrode system. Assuming a low frequency response to 0.1 Hz with 10-megohm input resistors, then 0.16 μ F coupling capacitors are required. The capacitive reactance of 0.16 μ F at 60 Hz is 16,700 ohms. When considering this impedance in the light of the earlier CMRR calculations, it becomes obvious that ordinary 10% tolerance capacitors will have a substantial impedance unbalance to this 60 Hz and thus will prevent a high common mode rejection ratio from being obtained. Even if the coupling capacitors are matched to $\pm 1\%$, the reduction in CMRR due to their imbalance will be significant. It is thus desirable not to use coupling capacitors at the inputs of a differential amplifier but to use them in later amplification stages after common mode signals have been rejected.

19.3 INPUT RESISTANCE

input
resistance
to common
mode and
differential
signals

For any differential amplifier there are two significant input resistance values: the input resistance of the amplifier to a differential signal and the input resistance to a common mode signal. From the preceding discussion it is clear that the input resistance to common mode signals (common mode input resistance) should be as high as possible, definitely above 10 megohms and preferably above 100 megohms. Relating back to the discussion on electrodes in Chapter 16, electrode offset potentials are stabilized over a period of time due to the loading effect of the amplifier; that is, the input resistance of the amplifier to differential signals will tend to discharge the "batteries" formed at the electrode-tissue interfaces. It is thus desirable that a differential amplifier's input resistance to differential signals (differential input resistance) be of a value that will allow electrode stabilization within two or three minutes after application. It is undesirable however, to use too low a differential input resistance as too low resistance produces a distorted pulse response to a physiological signal as discussed in Section 16.1.

In practice, a differential input resistance of 2 megohms has proved satisfactory and could perhaps be regarded as an optimum value for most physiological recording or from surface electrodes.

Referring to Fig. 19-4, the input resistance configuration of differential amplifiers determines the ratio between the differential input resistance and the common mode input resistance. The input resistance configuration shown in Fig. 19-4A uses separate input resistors to ground for each input.

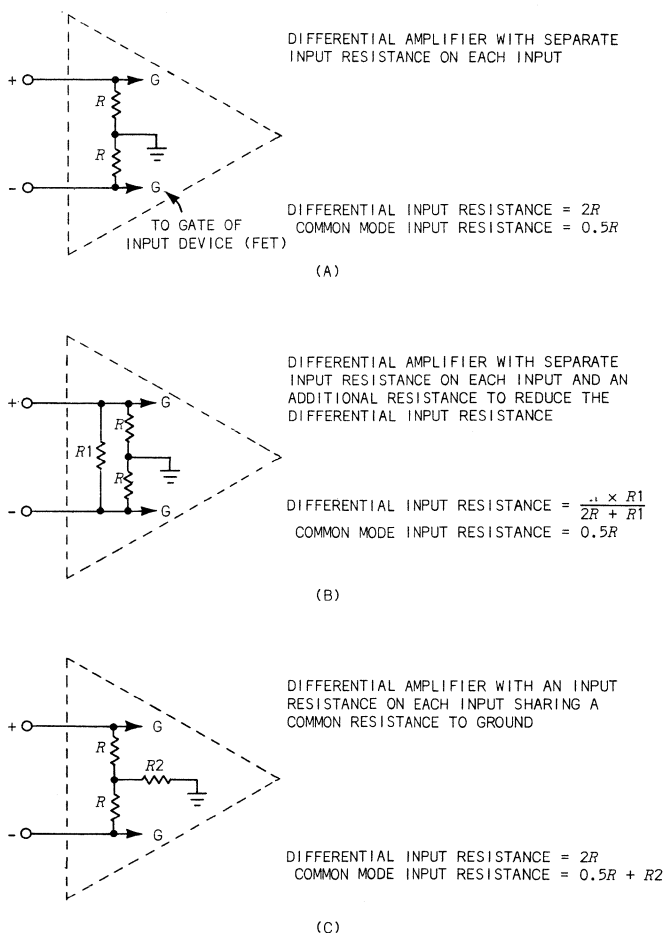
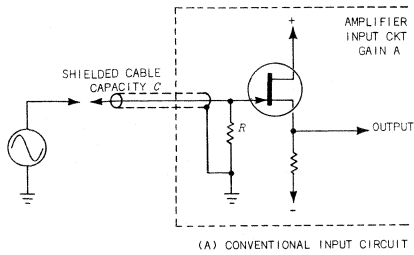


Fig. 19-4. Common mode and differential input resistance.

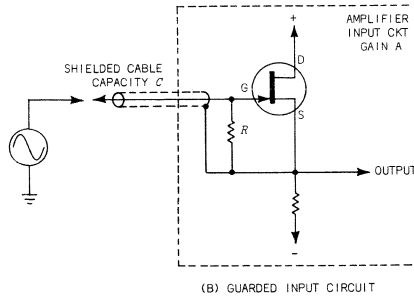
configuration determines ratio This configuration gives a high differential input resistance with a lower common mode input resistance, a somewhat undesirable situation. The input resistance configuration shown in Fig. 19-4B is similar to that shown in Fig. 19-4A with the addition of resistance R_1 to reduce the differential input resistance to an acceptable value. This configuration is satisfactory for use in most physiological situations and can be achieved from the circuit shown in Fig. 19-4A with the simple addition of one resistor. The input resistance configuration shown in Fig. 19-4C provides acceptable values for both the differential input resistance and the common mode input resistance and is particularly recommended as it has the added advantage of having only one physical resistor forming most of the common mode input resistance. Typical values for this circuit may be 1 megohm for R and 100 megohms for R_2 . This circuit also lends itself to input guarding as discussed in the following section.

19.4 INPUT GUARDING

single-ended input The common mode input impedance of an amplifier can be greatly increased by "guarding" or "bootstrapping" the input circuit of the amplifier. Input guarding can best be understood by viewing the amplifier from the input, by considering DC conditions and then expanding the analysis to include practical AC conditions. A conventional single-ended input circuit is shown in Fig. 19-5A. The input impedance of this circuit consists of the input resistance R in parallel with the reactance of the capacitance C of the shielded cable used for interconnection between the input voltage and the amplifier. Assume an input voltage of one volt and an input resistance R of one megohm. With the input voltage connected to the amplifier, a current of one μA will flow. As far as this input voltage is concerned, it is one volt seeing a load that requires one μA of current; the load thus appears as one megohm to the input voltage.

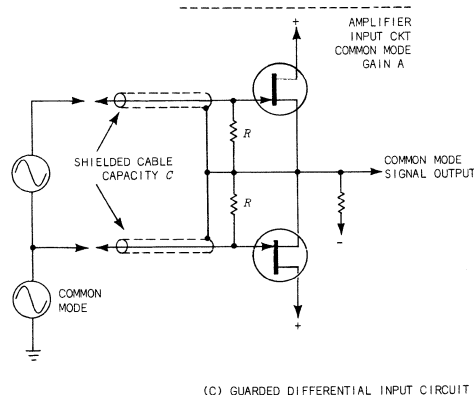


INPUT IMPEDANCE
 $Z_{in} = R \parallel X_C$



INPUT IMPEDANCE
 $Z_{in} = \frac{(R \parallel X_C)}{1 - A}$

TYPICAL VALUE FOR
 AMPLIFIER STAGE GAIN
 $A = 0.99$



COMMON MODE
 INPUT IMPEDANCE
 $Z_{in(cm)} = 0.5 \frac{(R \parallel X_C)}{1 - A}$

DIFFERENTIAL
 INPUT IMPEDANCE
 $Z_{in(dif)} = 2R$

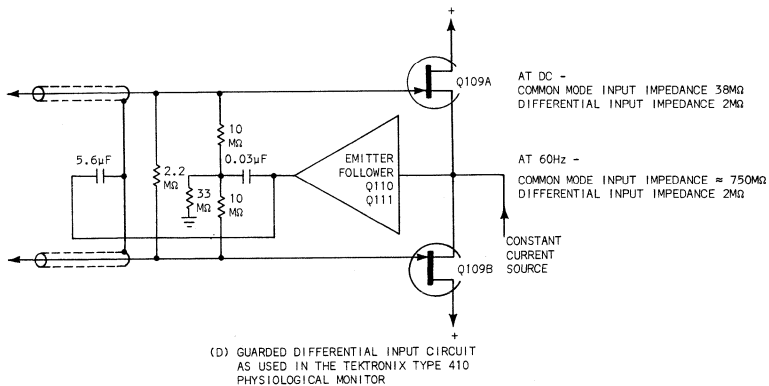


Fig. 19-5. Input guarding or "bootstrapping".

single-
ended
guarded
input

Now consider the single-ended guarded input circuit shown in Fig. 19-5B. Assume the same conditions as before with the input FET having a gain of 0.90 between gate and source. With the amplifier input resistance connected to the FET source rather than to ground, a one volt input will produce a one volt potential at the FET gate and therefore a 0.90 volt potential at the FET source. The potential therefore appearing across the input resistance R is the difference between the potentials, that is, 0.1 volt. The current through the one megohm input resistor is now 0.1 μ A. Now, as far as the input voltage is concerned, it is one volt seeing a load that requires 0.1 μ A of current; the load thus appears as 10 megohm to the input voltage even though the actual value of the input resistance is only one megohm. The value of the input resistance is thus amplified by a factor of 10 by connecting the input resistance in a guarded configuration rather than by simply connecting it to ground.

loop gain
amplifies
 R_{in}

In actual practice, a more typical value for the FET gain between the gate and the source would be 0.990, resulting in an input resistance "amplification" of 100X. Thus, a physical 1-megohm resistor in a guarded circuit would appear as a 100-megohm resistor to the input voltage. The closer the amplifier bootstrap loop gain is to unity, the higher the input resistance as seen from the input. The above discussion applies equally well to AC signals, however, both the input resistance R and the input capacitance C formed by the shielded cable connecting the input voltage to the amplifier must be considered as the input components of the circuit. As with the DC conditions discussed previously, the input reactance of the circuit can also be "amplified" by connecting the shield of the input cable to the FET source rather than simply to ground. A formula for the input impedance of this circuit is given in Fig. 19-5B.

common
mode
guarded
input

The above discussion can be further extended to a differential amplifier input circuit as shown in Fig. 19-5C. With a differential amplifier the common mode input impedance is increased by guarding. As discussed in the previous section, it is desirable to have as high a common mode input impedance as possible to minimize the effect of source impedance unbalance.

410 guarded
input

Tektronix Type 410 Physiological Monitor uses a guarded differential-input circuit as shown in Fig. 19-5D. Since high common mode input impedance is primarily required to reject 60-Hz signals, the guarding circuit can be AC coupled to the input resistors and cable shields as it is unnecessary to provide high common mode impedance to DC signals. The 410 Physiological Monitor has a 38-megohm common mode input resistance to DC signals and a common mode input impedance to AC signals of considerably above 750 megohms.

guarding
used within
amplifier

Guarding can also be used within an amplifier to provide constant operating conditions for the input FET's. In the 410 Physiological Monitor, as the gates of both input FET's go positive, the output of the input-guarding emitter follower, Q110 and Q111, also goes positive. This emitter-follower output is used to guard the various components of the input impedance and is also used to determine the drain potentials on each of the input FET's. The circuit is so arranged that a constant voltage appears between the source and drain no matter what the common-mode input-signal level.

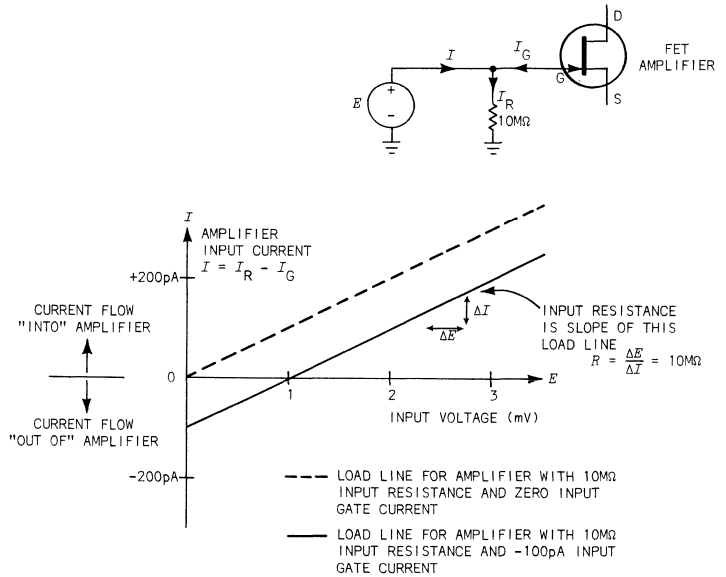


Fig. 19-6. Amplifier input current.

19.5 INPUT CURRENT

Consider the simplified amplifier configuration shown in Fig. 19-6. The input impedance of such an amplifier is defined as the ratio of an incremental change in the input voltage to the resulting incremental change in the input current. Thus, if the input voltage shown in Fig. 19-6 changed by 1 mV and the resulting input-current change was 100 pA, the input resistance would be $\frac{1 \text{ mV}}{100 \text{ pA}} = 10 \text{ megohm}$.

input gate
current

If the only component of input current was due to the input resistance, then Ohm's law dictates that the input current would be zero when the input voltage is zero. In all physiological amplifiers an additional input current, due to the amplification device, is always present. In the case of a modern FET amplifier, this input current would normally be below 1 nA and would possess a negative sign indicating that the FET is acting as a "source" of current rather than a "sink" of current. This current is referred to as "input gate current."

input grid current

In older style amplifiers using vacuum tubes or nuvistors for the input element, the input current may be several nanoamps and may be of either polarity, indicating that the input device is either a "sink" or "source" of current. Input current in these amplifiers is referred to as "input grid current."

input current versus input resistance

The two load lines shown on Fig. 19-6 defining the relationship between input voltage and input current both represent an input resistance of 10 megohm (the slope of the line), the upper line representing a *theoretically ideal* amplifier having zero input gate current and the lower line representing a *practical* FET amplifier having an input gate current of -100 pA . Since the component of the input current associated with the input amplification device is constant and does not vary with the input voltage, it must be shown separately when specifying an amplifier and cannot in any way be included with the amplifier's input resistance specifications.

input current produces offset potential

Input gate current can be a source of undesirable offset potential if an amplifier is used in conjunction with a high impedance source. Consider the amplifier shown in Fig. 19-6 used in conjunction with a source having a 1-megohm output impedance. Assume also that this source is zero volts. The -100 pA input grid current will flow through the output impedance of the source creating a potential drop of $100 \text{ }\mu\text{V}$. The output of the amplifier would then indicate a $100\text{-}\mu\text{V}$ input voltage even though the input voltage is in fact zero. This situation is only troublesome when making DC measurements as the gate current is a DC current.

input current compensation

Many amplifiers provide some degree of input gate-current compensation by adding a constant-current drain between the gates of the field effect transistors and ground. Thus, an amplifier such as the Tektronix Type 3A9 Differential Amplifier, has an input gate-current specification of $\pm 20 \text{ pA}$ at 25°C even though the actual internal FET-gate current may be in excess of -100 pA .

19.6 DYNAMIC RANGE, DC OFFSET AND RECOVERY

overloading
amplifiers

As either the common mode or differential input voltage to a differential amplifier is increased, a point is reached when the amplifier will overload and the output voltage will not be representative of the input voltage. This overload may occur within the amplifier input stages or elsewhere within the amplifier. If, however, the amplifier's dynamic range specifications are not exceeded, overload will not occur. Dynamic range characteristics for a differential amplifier are specified as follows:

1. Input differential dynamic range -- The maximum peak differential voltage that can be applied between the differential-amplifier input terminals,
2. Input common mode dynamic range -- The maximum peak common mode voltage that can be applied between the inputs of the differential amplifier and ground,
3. Output dynamic range -- The maximum peak output voltage that can be expected from the amplifier.

The amplifier is referred to as "overloaded" if any one of these three limits is exceeded.

transient
overloading
versus
recovery

In physiological applications it is not uncommon for transients to be produced that are several orders of magnitude greater in potential than the physiological signal under observation. An important characteristic of a physiological amplifier is the time taken for it to recover from such overloads. Amplifier recovery characteristics can be defined by the time taken for the amplifier to recover from a transient within the dynamic range specifications of the amplifier and by the time taken for the amplifier to recover from transients far greater than the dynamic range characteristics. DC-coupled amplifiers normally recover almost instantaneously from transients within the amplifier's dynamic range specifications as these transients are within the common mode rejection capabilities of the amplifier. Recovery time for transients that exceed the amplifier's dynamic range specifications are difficult to specify as

this recovery time depends on the precise characteristics of the overload transient. With such transients, thermal effects and the inherent unbalance produced by such thermal effects must be considered. Typical amplifiers may take several seconds to recover from severe overload.

AC coupling
and
transients
versus
recovery

AC-coupled differential amplifiers may have particularly bad recovery characteristics due to the inherent time constant involved in the AC-coupling network. Recovery time may be ten times as long as this time constant. Although AC coupling at the amplifier input is not recommended, AC coupling at a later stage in the amplifier is desirable as it eliminates electrode offset potentials, etc. Amplifiers that are AC coupled after several stages of DC-coupled amplification are referred to as "AC-stabilized" amplifiers. Many physiological amplifiers, such as the amplifier incorporated into the Tektronix Type 410 Physiological Monitor, incorporate AC coupling after several stages of DC amplification and internal limiting to improve the amplifier's recovery characteristics. The amplifier in the Type 410 Physiological Monitor has a DC-coupled input stage and is AC coupled after this input stage at a time constant of two seconds. Its recovery time is less than four seconds for any overload signal. This characteristic is particularly important in an amplifier intended for ECG monitoring in intensive care and surgery as the amplifier will certainly be overloaded if the subject is defibrillated and it is highly important that the medical personnel be able to use the amplifier immediately after the defibrillation process is complete.

410 recovery

DC
differential
offset
versus
electrode
offset

Physiological signals obtained from electrodes include a DC component due to the electrode-offset potential. A differential amplifier capable of recording these physiological signals in a DC-coupled mode must therefore, include the capability to reject this electrode-offset potential. This is achieved by incorporating DC-differential offset within the differential amplifier which effectively produces a DC-differential voltage at the differential amplifier input which cancels the electrode-offset potential.

electrode offset exceeding dynamic range

If DC-differential offset is not available within an amplifier, AC coupling must be used to reject the electrode-offset potential. As it is undesirable to incorporate AC coupling at the amplifier's input the amplifier must be AC coupled after one or more amplification stages and care must be taken to ensure that the electrode-offset potential is not exceeding the dynamic-range characteristics of the amplifier. Often the user may not be aware that the amplifier's dynamic range is being exceeded and erroneous results are obtained.

dynamic window

The term "dynamic window" refers to the use of a differential amplifier with DC-differential offset. Dynamic window can best be understood by considering the following example. Consider a Tektronix Type 3A9 Differential Amplifier operating at its maximum gain of 100,000 with no DC-differential offset. As the output dynamic range for this amplifier is ± 5 volts, the maximum differential input voltage that will not exceed this output dynamic range is $\frac{\pm 5 \text{ volts}}{100,000}$ or $\pm 50 \mu\text{V}$. The amplifier is able to "see" a dynamic window of $\pm 50 \mu\text{V}$. Now assume that the Type 3A9 is used in the DC OFFSET position. The DC OFFSET range available is the same as the differential dynamic range, that is ± 1 volt. Assume that the course and fine DC OFFSET controls are set to provide $+0.3$ volts of DC-differential offset. Now the maximum differential input voltage that can be applied to the amplifier is $\pm 50 \mu\text{V} + 0.3$ volts. The amplifier is now able to "see" a dynamic window of $\pm 50 \mu\text{V}$ from 0.3 volts, that is from 0.299950 volts to 0.300050 volts.

19.7 NOISE AND DRIFT

noise versus drift

Noise and drift are both unwanted signals that occur within an amplifier. The term "noise" normally refers to unwanted signals generated at a frequency above 0.1 Hz and DC drift normally refers to slow changes in the output level at frequencies below 0.1 Hz. The term "stability" is also used to indicate relative DC-drift characteristics. The component of noise occurring at the power-line frequency and harmonics of the power-line frequency is often referred to as "hum."

stability

hum

qualifying
noise

The noise generated within an amplifier is often specified in microvolts and refers to this noise as if it were a differential input voltage. The noise produced within an amplifier is either specified as microvolts P-P, microvolts RMS, microvolts tangential or by specifying an equivalent noise resistance. The equivalent noise resistance considers all the noise in an amplifier to be produced by an equivalent resistance producing thermal noise. The RMS value of this thermal noise is related to the resistance value by the following expression:

$$E_{\text{RMS}} = 7.4 \times 10^{-12} \times \sqrt{R \text{ bw } (t + 273)}$$

or

equivalent
related to
RMS

$$R = \frac{(E_{\text{RMS}})^2 \times 10^{24}}{55 \text{ bw } (t + 273)}$$

R = Equivalent Noise Resistance -- ohms
 bw = Bandwidth -- Hz
 t = Temperature -- °C

Thus, an amplifier specified as having an equivalent noise resistance of 100,000 ohms would produce 7 μV RMS of noise at 30 kHz bandwidth at 20°C. Equivalent noise resistance relates, therefore, to RMS noise.

RMS noise

The RMS noise, and thus the equivalent noise resistance, of an amplifier is measured by shorting the inputs to ground and measuring the output voltage at maximum sensitivity with an RMS-indicating voltmeter. The RMS noise generated within the amplifier will then be this RMS output voltage divided by the gain of the amplifier. For a Tektronix Type 3A9 Differential Amplifier operating at a bandwidth of 30 kHz and at its maximum gain of 100,000, the RMS noise output was 0.32 volts. Thus, the RMS noise produced within this amplifier is 3.2 microvolts RMS and the equivalent noise resistance for the amplifier is 20,000 ohms.

Amplifier noise is normally specified in microvolts RMS if the amplifier is to form part of an instrumentation system. This noise specification gives no information as to the peak component of noise or to the relative appearance of a CRT display incorporating this noise, thus either P-P noise or tangential-noise measurements may be desirable in some instances.

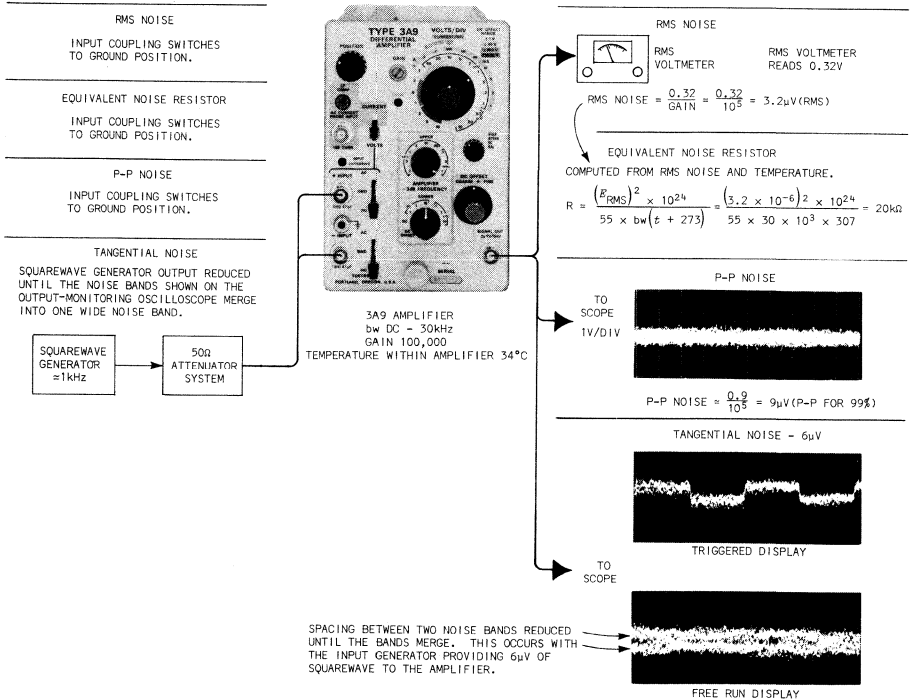


Fig. 19-7. Amplifier noise measurement.

peak-to-
peak noise

If an amplifier is to be used in conjunction with peak detector or trigger circuitry, it is important to know the peak value of noise produced by an amplifier rather than simply the RMS value of this noise. Peak-to-peak noise for an amplifier can be measured in the same way as RMS noise by using an oscilloscope to display the noise output. The displayed noise for the amplifier system discussed previously is shown in Fig. 19-7. From observation of the oscilloscope display at a sensitivity of 1 volt per division, the P-P deflection shown appears to be approximately 0.9 divisions referring to an output of 0.9 volts. The P-P noise generated within the amplifier is thus $9 \mu\text{V}$ P-P referred to the input. Closer examination of the noise shows that, while 99 percent of the information appears to be within 0.9 divisions, random spikes are produced over an apparent overall range of 1.2 divisions. (Theoretically, random spikes can extend out any distance.) The P-P noise generated by this amplifier could, therefore, be more correctly stated as $12 \mu\text{V}$ subjective P-P with 99 percent of the noise below $9 \mu\text{V}$ P-P.

tangential
noise

In the majority of physiological measurements an oscilloscope is used to display the output from an amplifier. It is thus desirable to know the amplifier's noise characteristics with respect to this oscilloscope display. A "tangential" measurement of displayed noise effectively measures the trace width produced by noise and therefore gives an idea of the resolution obtainable from the oscilloscope screen. The following discussion on tangential noise measurement technique will further clarify this resolution concept. Consider a low-amplitude squarewave being applied to the input of the differential amplifier as shown in Fig. 19-7. As it is necessary to know the amplitude of this squarewave, it is desirable to obtain a relatively high-amplitude squarewave for ease of measurement and then to accurately attenuate this via a 50-ohm attenuation system. Although the output from this 50-ohm attenuation system is single ended, that is, only one output source is produced with respect to ground, this output should be coupled differentially into an amplifier by connecting the positive input of the amplifier to the output source and the negative input of the amplifier to the reference ground for this output source. For the configuration

shown in Fig. 19-7, the oscilloscope will clearly show the squarewave together with the noise generated from within the differential amplifier; this display is shown in the upper photograph of the tangential noise section of Fig. 19-7. If the oscilloscope's sweep generator is then operated in a free-running mode, the squarewave information will be lost and the CRT display will appear as two bands of noise. As the output from the squarewave generator is increased in amplitude, these bands will tend to separate; as the output is decreased in amplitude, these bands will tend to merge into one. The input squarewave level at which these two bands just merge into a single band is referred to as the tangential-noise specification for the amplifier. In the particular configuration shown in Fig. 19-7, the tangential noise specification for the amplifier is $6 \mu\text{V}$ and a free-running display of this $6\text{-}\mu\text{V}$ squarewave together with inherent noise from within the amplifier is shown in the bottom photograph in Fig. 19-7.

tangential
 $\times 2$ equals
 RMS

If an amplifier's frequency response is perfectly Gaussian, then a mathematical analysis of RMS-noise measuring techniques and tangential-noise measuring techniques will show tangential noise to be twice RMS noise. Since most physiological amplifiers are essentially Gaussian, RMS noise can be simply determined once tangential noise has been measured. This indirect method of determining RMS noise sidesteps the need for a true RMS-reading voltmeter.

P-P
 derivation
 from RMS

If the distribution of noise within an amplifier is Gaussian, and the noise distribution in most wideband amplifiers is approximately Gaussian, then the noise distribution will follow a "normal distribution" curve. From the curve we can deduce that:

RMS noise $\times 2$ = P-P noise for 68% of the noise
 RMS noise $\times 4$ = P-P noise for 96% of the noise
 RMS noise $\times 5$ = P-P noise for 99% of the noise
 RMS noise $\times 6$ = P-P noise for 99.9% of the noise
 RMS noise $\times 8$ = P-P noise for 99.99% of the noise

Examination of the equation relating RMS noise to equivalent noise resistance indicates that RMS noise is proportional to the square root of the amplifier's bandwidth. This noise component is known as "thermal noise" or "Johnson noise." The above proportionality is essentially true for wideband amplifiers (100 kHz and above) but, however, does not hold true for narrowband amplifiers or when wideband amplifiers are subject to bandwidth limiting. Under these conditions an additional low-frequency noise component generated within the input field-effect transistors becomes an appreciable proportion of the amplifier's total noise and cannot be neglected. This extra noise component is referred to as "excess noise" (also occasionally referred to as " $\frac{1}{f}$ noise" or "flicker noise") and is discussed in more detail in Chapter 20 when discussing bandwidth-limiting considerations.

excess
noise

As previously stated, drift can be regarded as very low-frequency noise generated within an amplifier. Drift specifications, as with noise specifications, are in microvolts and refer to the drift as if it were a differential input voltage. Because of its inherent low frequency nature, drift is normally specified in P-P units. Drift specifications for most amplifiers are self-explanatory, however, a full evaluation of drift should include both long-term drift and short-term drift measurements. Short-term drift measurements are measured in microvolts per minute and indicate to the user the drift in microvolts expected over a one-minute period after the amplifier has had sufficient time to warm up. Long-term drift specifications are in microvolts per hour and indicate the total drift that may be expected over a one-hour period. Long-term drift specifications will always exceed short-term drift specifications.

short- and
long-term
drift

Both long-term drift and short-term drift assume a constant operating temperature within the differential amplifier and an additional specification must be generated to determine the characteristics of this drift with a change in temperature. Temperature drift is specified in microvolts per degree Celsius and indicates the drift that may be expected from the amplifier for a 1°C change in ambient temperature.

temperature
drift

PRINCIPAL CHARACTERISTICS ONLY SHOWN WITH THE 3A9 USED AS AN INPUT AMPLIFIER TO PROVIDE GAIN FROM A DIFFERENTIAL INPUT TO THE SIGNAL OUTPUT CONNECTOR AT A BANDWIDTH OF FROM DC TO 30kHz. BOTH WIDER AND NARROWER BANDWIDTHS ARE AVAILABLE AS DISCUSSED IN CHAPTER 20 - SIGNAL PROCESSORS. GAIN SETTINGS BELOW $\times 100$ ARE NOT CONSIDERED.

CHARACTERISTICS	PERFORMANCE	* MODIFIED PERFORMANCE
GAIN - CALIBRATED	$\times 100$ TO $\times 100,000$; 10 STEPS IN A 1-2-5 SEQUENCE	SAME AS STANDARD PERFORMANCE
- UNCALIBRATED	VARIABLE CONTROL PROVIDES CONTINUOUSLY VARIABLE GAIN BETWEEN STEPS	"
CMRR	100,000:1	"
DYNAMIC RANGE	± 1 VOLT DIFFERENTIAL ± 10 VOLTS COMMON MODE	"
RECOVERY	TO WITHIN 0.5% IN LESS THAN 10 μ s IF OVERLOAD VOLTAGE WITHIN DYNAMIC RANGES SHOWN ABOVE	"
INPUT IMPEDANCE	2M Ω 24pF DIFFERENTIAL 0.5M Ω 94pF COMMON MODE	5.0M Ω 94pF COMMON MODE
INPUT CURRENT	MAXIMUM ± 20 pA AT EACH INPUT AT 25 $^{\circ}$ C	TYPICALLY $< \pm 100$ pA AT EACH INPUT AT 25 $^{\circ}$ C
NOISE	6 μ V MEASURED TANGENTIALLY (TYPICAL)	SAME AS STANDARD PERFORMANCE
DC DRIFT AFTER 1 HR WARMUP	5 μ V/MIN (PEAK TO PEAK) 10 μ V/HOUR (PEAK TO PEAK) 50 μ V/ $^{\circ}$ C	"
OUTPUT DYNAMIC RANGE	± 5 VOLTS FROM 100 Ω SOURCE	"
DC OFFSET	± 1 VOLT	"

*3A9 MODIFIED TO INCREASE INPUT IMPEDANCE
- REMOVE LINKS ON 3A9 CIRCUIT BOARD AS SHOWN ON PHOTOGRAPH.
- ADD 2.2M Ω RESISTOR BETWEEN + INPUT AND - INPUT IF DESIRED TO MAINTAIN 2M Ω DIFFERENTIAL INPUT RESISTANCE.

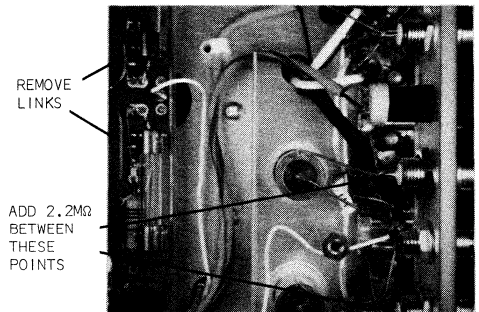


Fig. 19-8. Characteristics of the Tektronix Type 3A9 Differential Amplifier when used as a physiological differential amplifier.

power
supply
drift

If high-stability power supplies are not used within an amplifier, drift may be caused by variations in supply voltage. This would be specified as microvolts drift for a supply voltage change of $\pm 10\%$ from nominal. Drift with line-voltage change is negligible in modern FET amplifiers.

19.8 SPECIALIZED AMPLIFIERS

optimum
character-
istics

Only the characteristics of general-purpose physiological amplifiers have been discussed in this chapter. A typical general-purpose physiological amplifier is the Tektronix Type 3A9 Differential Amplifier, abbreviated specifications for which are shown in Fig. 19-8. Many physiological-monitoring situations are particularly dependent on one or more characteristics of an amplifier and it is often necessary to use amplifiers with characteristics optimized for a particular measurement requirement. An ECG amplifier intended for use during defibrillation must possess particularly good recovery characteristics. An EEG amplifier must be particularly free of noise and an amplifier intended for use with microelectrodes for recording evoked responses must have an extremely high input impedance and low input current. Amplifiers specifically intended for use with microelectrodes are discussed in more detail in Chapter 11. A relatively new concept is the use of plug-in probes that allow a physiological amplifier to be optimized for a particular measurement requirement. Thus, the one amplifier may be used for general purpose recording, for high sensitivity recording, or for use in conjunction with microelectrodes simply by changing probes.

20

SIGNAL PROCESSORS – OPERATIONAL AMPLIFIERS

The signal available as an output from a physiological amplifier may not be in a form compatible with other instrumentation. Signal processing is often required to either achieve compatibility or to improve the presentation of a physiological signal. Signal processing is also used when more than one category of information is to be obtained from the same physiological signal.

20.1 OPTIMUM BANDWIDTH FOR PHYSIOLOGICAL SIGNALS

DC-10,000 Hz
bandwidth

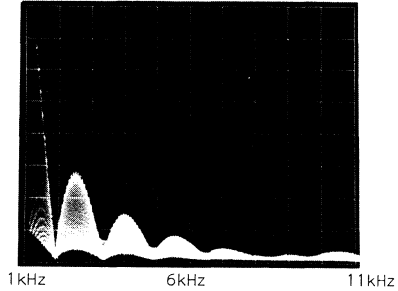
Spectrum analysis of the action potential generated when a single cell depolarizes indicates that essentially all of the signal is produced within the frequency domain from DC to 10,000 Hz. Since all physiological signals are the sum of one or more action potentials, the frequency domain of these physiological signals will be less than or equal to the frequency domain of the single-cell action potential. Thus, the optimum bandwidth for a physiological amplifier capable of monitoring any physiological signal is from DC to 10,000 Hz. This bandwidth is adequate for use in almost all physiological monitoring situations; however, in the research environment, some workers postulate the existence of higher frequency components and a bandwidth from DC to 30,000 Hz should be adequate to allow these postulated components to be observed.

INPUT SIGNAL

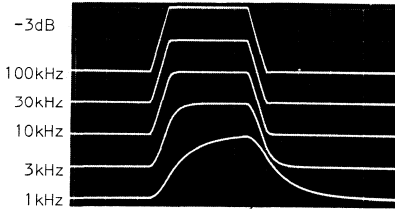
PRF = 100Hz
 t_r AND t_f = 0.1ms
 WIDTH = 0.6ms



FREQUENCY SPECTRUM OF INPUT SIGNAL



HIGH FREQUENCY RESPONSE LIMITING



DISPLAYED RISE-
TIME OR FALLTIME

$$= \sqrt{\left(\frac{\text{MAXIMUM RISE- OR FALLTIME COMPONENT IN THE INPUT SIGNAL}}{\text{HIGH FREQUENCY -3dB POINT}} \right)^2 + \left(\frac{0.35}{\text{HIGH FREQUENCY -3dB POINT}} \right)^2}$$

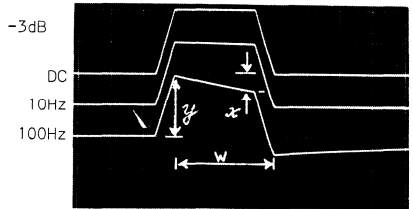
HIGH FREQUENCY -3dB CUTOFF REQUIRED FOR NEGLIGIBLE DISTORTION, THAT IS, NO VISUAL WAVEFORM CHANGE

$$= \frac{1.5}{\text{MAXIMUM RISE- OR FALLTIME COMPONENT IN THE INPUT SIGNAL}}$$

HIGH FREQUENCY -3dB CUTOFF REQUIRED FOR ACCEPTABLE DISTORTION, THAT IS, NO VISUAL RISE TIME DEGRADATION BUT SOME ROUNDING OF ABRUPT CHANGES

$$= \frac{1.0}{\text{MAXIMUM RISE- OR FALLTIME COMPONENT IN THE INPUT SIGNAL}}$$

LOW FREQUENCY RESPONSE LIMITING



DISPLAYED "TILT" OR "SAG" IN % = $\frac{x}{y} \times 100\%$
 DISPLAYED TILT

$$= \% \text{ TILT ON THE INPUT SIGNAL} + \left(6.30 \times w \times \text{LOW FREQUENCY -3dB POINT} \right)$$

LOW FREQUENCY -3dB CUTOFF REQUIRED FOR NEGLIGIBLE DISTORTION, THAT IS, < 1% TILT ADDED

$$= \frac{.0016}{\text{MAXIMUM } w \text{ IN THE INPUT SIGNAL}}$$

LOW FREQUENCY -3dB CUTOFF REQUIRED FOR ACCEPTABLE DISTORTION, THAT IS, < 5% TILT ADDED

$$= \frac{.008}{\text{MAXIMUM } w \text{ IN THE INPUT SIGNAL}}$$

Fig. 20-1. Input signal degradation due to frequency response limiting.

distortion
versus
bandwidth-
risetime
product

The optimum bandwidth requirement of 10,000 Hz proposed in the previous paragraph is somewhat theoretical as it supposes that a single-cell action potential could be coupled to an amplifier without distortion. In practice, these potentials are distorted by the monitoring techniques used and a bandwidth of substantially less than 10,000 Hz will not produce additional distortion. No visual distortion of most waveforms can be observed on an oscilloscope if the product of the oscilloscope's amplifier bandwidth and the fastest risetime component in the signal exceeds 1.5. Only very slight distortion can be observed on an oscilloscope if this bandwidth-risetime product is between 1 and 1.5. These relationships should thus be observed when considering the optimum bandwidth required for monitoring a particular physiological signal.

simulated
action
potential

Referring to Fig. 20-1, a single-cell action potential can be experimentally simulated with a pulse generator having a rise- and falltime of 0.1 ms and a pulse width of 0.6 ms. Such a pulse is shown in Fig. 20-1 together with a frequency domain presentation showing the predominant energy content of this pulse to be below 10,000 Hz. The effects on this pulse of both high-frequency response limiting and low-frequency response limiting are also shown in Fig. 20-1.

20.2 AMPLIFIER NOISE REDUCTION BY BANDWIDTH LIMITING

The noise generated within a physiological amplifier is discussed in Chapter 19, Section 19.7. This discussion refers to two sources of noise within an amplifier, "Johnson noise" and "excess noise."

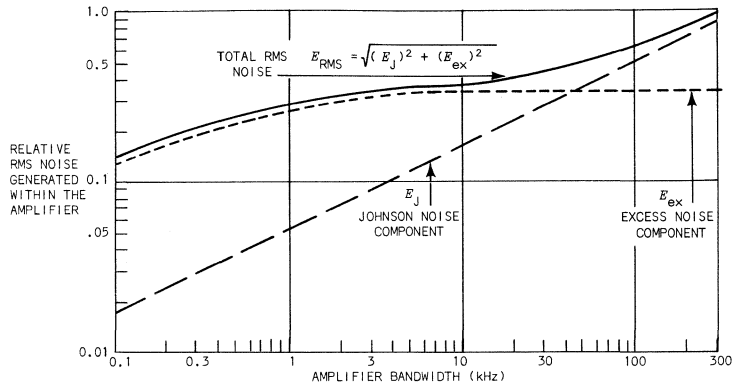


Fig. 20-2. Amplifier noise/bandwidth relationship (Tektronix Type 3A9 Amplifier).

Johnson
noise
versus
bandwidth

Examination of the equation relating the Johnson noise produced within an amplifier to the amplifier's equivalent noise resistance indicates that Johnson noise is proportional to the square root of the amplifier's bandwidth. It is thus apparent that if the amplifier's bandwidth is limited, the RMS noise produced within the amplifier will also be decreased. This Johnson noise-bandwidth relationship is shown in Fig. 20-2.

excess
noise
versus
bandwidth

The excess noise component of the total RMS noise produced within an amplifier is not mathematically related to the amplifier's bandwidth. There is little change in this excess noise component when the amplifier's bandwidth is reduced from 300,000 Hz to 3,000 Hz, thus the frequency content of this noise for modern FET amplifiers is predominantly below 3,000 Hz. The relationship between the excess noise produced within a Tektronix Type 3A9 Amplifier and the amplifier's bandwidth is also shown in Fig. 20-2. Excess noise has historically been referred to as $\frac{1}{f}$ noise.

total noise
versus
bandwidth

Since both Johnson noise and excess noise are essentially random, the total of these two noise components will be the square root of the sum of the squares of each component. The total RMS noise produced within a Type 3A9 Amplifier is shown in Fig. 20-2. It is apparent from this figure that an amplifier bandwidth reduction from 300,000 Hz to 10,000 Hz results in a 280% reduction in the noise produced within the amplifier. It is also apparent that a further reduction in the amplifier's bandwidth from 10,000 Hz to 1,000 Hz is of less value in reducing noise as the resulting decrease in the amplifier's noise is only about 33%. Many physiological signals can be recorded with bandwidths substantially less than 1,000 Hz and in such cases, limiting the amplifier's frequency response can also substantially reduce the noise generated within the amplifier.

3A9 signal
processing
features

An amplifier such as the Tektronix Type 3A9 Amplifier incorporates bandwidth limiting as shown on Fig. 20-3. The high frequency response of the amplifier can be adjusted in nine steps between 1 MHz and 100 Hz and the low frequency response of the amplifier can be adjusted in seven steps between DC and 10,000 Hz. DC differential-offset capabilities are also provided to offset the effect of electrode-tissue interface potentials as discussed in Chapters 16 and 19.

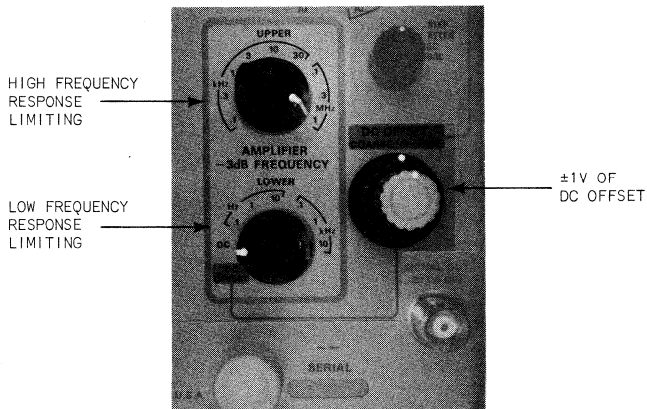


Fig. 20-3. Signal processing controls on the Type 3A9 Amplifier.

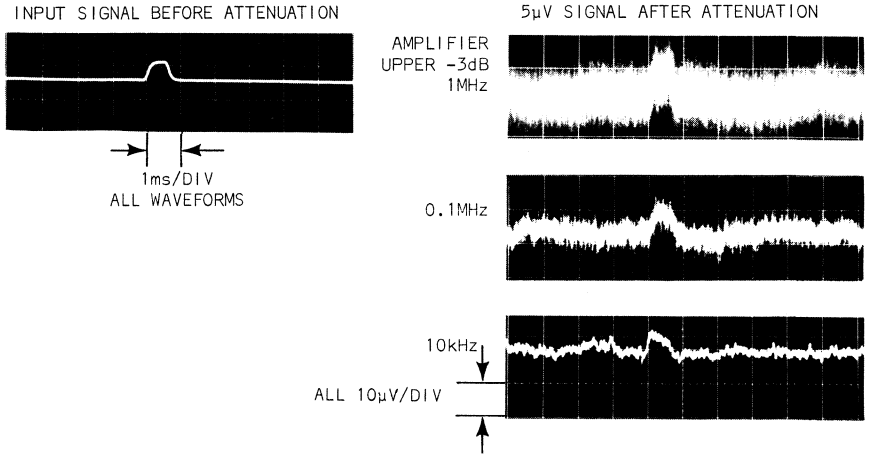


Fig. 20-4. Noise reduction by high frequency response limiting.

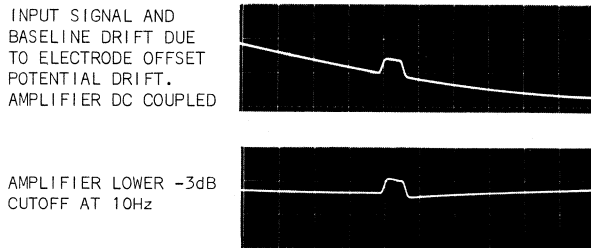


Fig. 20-5. Drift reduction by low frequency response limiting.

limiting
upper 3-dB
point

The effect of bandwidth limiting within an oscilloscope vertical amplifier on the oscilloscope's display is shown in Fig. 20-4. With the oscilloscope displaying a 5- μ V pulse with an amplifier bandwidth of 1 MHz, the pulse is completely swamped by the amplifier's internally generated noise. If the amplifier's upper 3-dB cutoff is limited to 10,000 Hz the noise is substantially reduced and the input signal predominates. A further reduction in the amplifier's bandwidth has little effect on the displayed noise; however, it does produce undesirable distortion.

20.3 AMPLIFIER LOW FREQUENCY RESPONSE LIMITING

limiting
lower 3-dB
point

While limiting an amplifier's low frequency response has a negligible effect on the noise produced within the amplifier, it can be useful in eliminating unwanted low frequency signals such as drift within the amplifier or drift in the electrode-offset potential produced between two physiological electrodes. Fig. 20-5 depicts a typical physiological response obtained with a DC-coupled amplifier and one obtained using low frequency bandwidth limiting.

20.4 LINE FREQUENCY REJECTION

Some years ago line-frequency rejection filters were incorporated into physiological monitoring systems to eliminate 50- or 60-Hz line frequency interference from a physiological signal. Of course, if the physiological signal also contained frequency components at 50 or 60 Hz, these components were also eliminated and the physiological signal was distorted.

high CMRR
minimizes
interference

With the advent of the modern high CMRR differential amplifier it was found that the increased common mode rejection eliminated most line-frequency interference, indicating that this interference was from a common mode source rather than from a differential source. When using an amplifier with high common mode rejection ratio, it is rarely necessary to include any form of line frequency filter and, as such filters may cause signal distortion, line-frequency rejection filters should be considered only as a last resort.

20.5 NOISE REDUCTION BY SIGNAL AVERAGING

signal averaging minimizes noise

Signal averaging is a signal processing technique for extracting a wanted signal from a background of unwanted noise. Signal averaging can only be used if the desired signal can be generated a number of times, either periodically or aperiodically, and can be used to effectively reduce the noise accompanying the signal by several orders of magnitude. The subject of signal averaging is adequately covered in Chapter 11, Section 11.4, when discussing the techniques used to record evoked cortical responses. Because of its inherent expense, signal averaging is normally used only when absolutely necessary and is primarily used for recording cortical responses and in research applications. It is, however, an important physiological signal processing technique and should not be overlooked. This subject is discussed elsewhere in this book.

20.6 OPERATIONAL AMPLIFIERS

characteristics

Electronically, an operational amplifier is simply a high gain inverting amplifier designed to remain stable with large amounts of negative feedback from output to input. An operational amplifier, by means of this negative feedback, is capable of processing a signal in many different ways, dependent only on the passive components selected as the feedback network and possibly as a series input network. The ideal operational amplifier has infinite gain, infinite input impedance and zero output impedance.

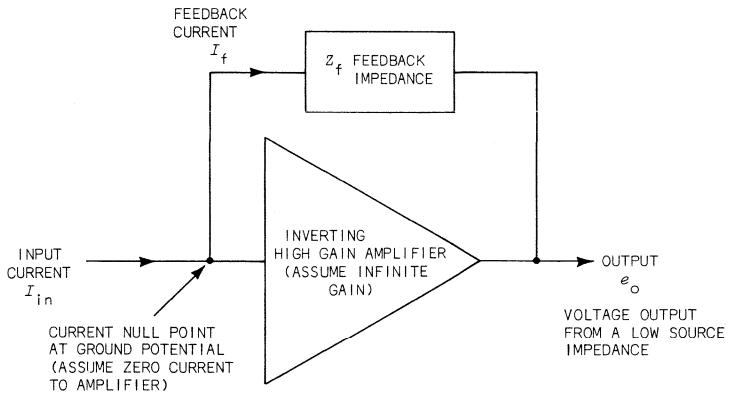
reference sources

This chapter should not be regarded as a general reference on operational amplifiers, it specifically discusses operational amplifiers from a biophysical measurement standpoint and makes no attempt to review some of the more sophisticated operational amplifier systems used within the general electronics industry. A companion volume entitled *Operational Amplifier Circuits* published by Tektronix, Inc. provides a more general reference source on the use of operational amplifiers. The instruction manual provided by Tektronix for use with the Type 3A8 Operational Amplifier plug-in also provides an extensive reference on operational-amplifier signal-processing circuits. The major manufacturers of modular or IC-packaged operational amplifiers also produce

application handbooks containing many circuits that would be useful for biophysical measurements. In particular, we would recommend *Applications Manual for Operational Amplifiers* published by Philbrick/Nixus Research, Dedman, Massachusetts.

negative
feedback to
current
null at
input

Referring to Fig. 20-6, an operational amplifier with negative feedback operates in a self-balancing configuration providing, through the feedback element, whatever current is necessary to hold the input at ground potential. The output signal is a function of this current and of the impedance of the feedback element. If current is applied to the operational amplifier input, it would tend to develop a voltage across the feedback impedance and thus move the operational amplifier input away from ground potential. The output, however, changes in the opposite direction, providing current to balance the input current and thus maintains the operational amplifier input at ground potential. If the feedback impedance is high, the output voltage must become quite high to provide enough current to balance even a small input current.



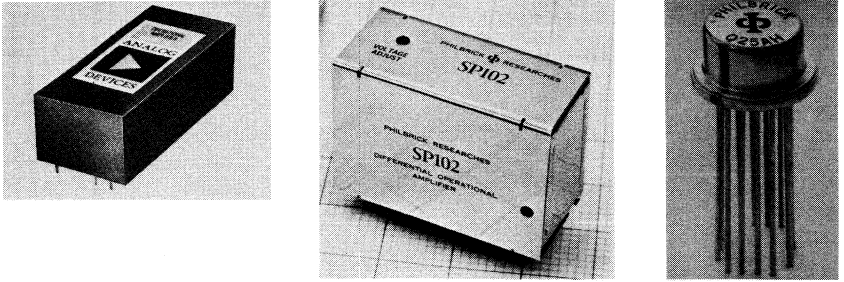
$$I_{in} + I_f = 0$$

$$\text{AND } I_f = \frac{e_o}{Z_f}$$

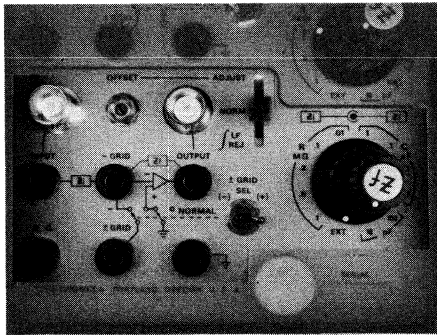
$$\therefore I_{in} + \frac{e_o}{Z_f} = 0$$

$$\text{OR } e_o = -\frac{I_{in}}{Z_f}$$

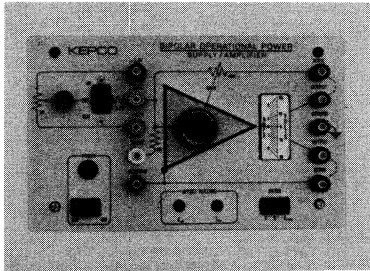
Fig. 20-6. Operational amplifier principle.



(A) MODULAR OPERATIONAL AMPLIFIERS USING DISCRETE COMPONENTS OR IC'S
GENERALLY LOW POWER OUTPUTS
<100mW



(B) TEKTRONIX TYPE 3A8 OPERATIONAL AMPLIFIER PLUG-IN UNIT
MEDIUM POWER OUTPUT
 $\pm 25V$ $\pm 7.5mA$ 190mW



(C) KEPKO POWER OPERATIONAL AMPLIFIER
HIGH POWER OUTPUT
 $\pm 36V$ $\pm 5A$ 180W

Fig. 20-7. Typical operational amplifiers.

voltage
input

In physiological measurements we are more often required to deal with an input *voltage* rather than an input *current*. We thus use an additional element, referred to as an input impedance (Z_i), which converts an input voltage signal to an input current signal.

noninverting
input

The input to an inverting high-gain amplifier is commonly referred to as the inverting input or "- input." Many operational amplifiers also provide access to a noninverting or "+ input." If this noninverting input is used in an operational amplifier circuit, negative feedback must still be maintained between the output and inverting input and the operating principles remain as discussed previously with the exception that the current null point will be maintained at the potential of the noninverting input rather than at ground potential.

classifi-
cations

Many types of operational amplifiers are commercially available, some of which are shown in Fig. 20-7. Operational amplifiers can generally be classified as either general purpose or specialized, this discussion will refer only to general purpose operational amplifiers. Operational amplifiers can be further classified as to their output voltage and current capabilities.

low power

Low power operational amplifiers are available in modular or integrated circuit packages as shown in Fig. 20-7A and generally provide an output power of less than 100 milliwatts. A typical IC-packaged low-power operational amplifier may provide a maximum output of ± 10 volts at ± 5 milliamperes. Modular operational amplifiers are particularly suited for use in custom signal processing equipment as they need only be supplied with the necessary DC power, feedback impedance and input impedance to perform a signal processing function. The Fairchild $\mu A741$ operational amplifier is particularly suitable.

medium
power

The operational amplifier shown in Fig. 20-7B is one of the two operational amplifiers incorporated into the Tektronix Type 3A8 Operational Amplifier plug-in unit. The Type 3A8 can be used with Tektronix 560-Series Oscilloscopes or with the Type 129 Plug-in Unit Power Supply. It contains two separate but identical operational amplifiers plus a display amplifier. The display amplifier monitors the output of either operational amplifier or can be used as an independent oscilloscope amplifier. The input impedance and feedback impedance for each operational amplifier can be internally selected from a range of resistors and capacitors, thus eliminating the need for external components if the operational amplifier is to be used for simple signal processing applications. If more sophisticated applications are required the necessary input impedance and feedback impedance networks can be permanently constructed on an adapter assembly (Tektronix PN 013-0048-01) which can be plugged into the operational amplifier when desired. This technique is used extensively in the various operational amplifier circuits used throughout this book and discussed in greater detail in Chapter 29. The operational amplifiers incorporated into the Type 3A8 are classified as medium-power operational amplifiers providing an output power below 1 watt. The Type 3A8 can provide a maximum output of ± 25 V at ± 7.5 mA.

high power

The high-power operational amplifier shown in Fig. 20-7C has an output power capability of 180 watts. Such high power capability operational amplifiers are rarely used in biophysical measurements; however, high-power operational amplifiers having output powers of several watts may be required to drive chart recorders and other low impedance devices.

20.7 OPERATIONAL AMPLIFIER APPLICATIONS

The references mentioned at the beginning of the previous section (20.6) should be consulted when considering the use of an operational amplifier for a particular signal processing application. Operational amplifier applications can, however, be classified into several broad groups as shown in Figs. 20-8 and 20-9. The following will discuss

these configurations as relating to one of the operational amplifiers in the Tektronix Type 3A8 Operational Amplifier plug-in unit. Some practical operational amplifier circuits using the Type 3A8 are discussed in Chapter 29.

noninverting amplifier

The noninverting amplifier shown in Fig. 20-8 can be used to provide gain in any physiological measuring system. It also provides a high input impedance and a low output impedance and thus can provide current gain as well as voltage gain in a system. If the Z_i and Z_f components of this noninverting amplifier are infinity and zero respectively, then the configuration becomes a unit voltage gain amplifier providing only current gain between the input and output.

oscillator

If positive feedback is added to an operational amplifier, an oscillator is formed as shown in Fig. 20-8. One or more frequency dependent elements in the positive-feedback circuit determines the frequency of oscillation, however, some negative feedback is also required to stabilize the oscillation amplitude. This negative feedback may simply consist of resistive feedback elements or often nonlinear devices are used to provide some degree of output voltage regulation.

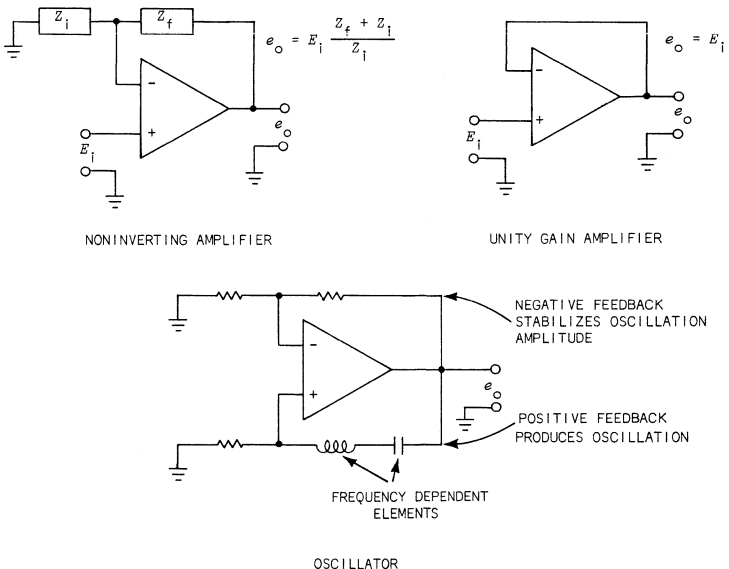
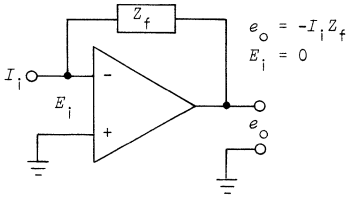
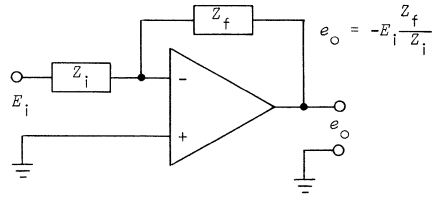


Fig. 20-8. Operational amplifier applications using the noninverting (+) input to the amplifier



CURRENT TO VOLTAGE CONVERTER

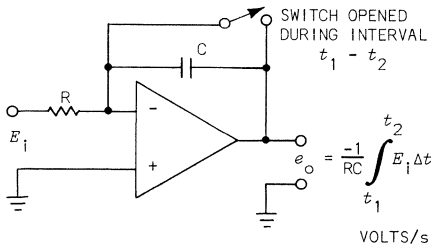


INVERTING AMPLIFIER

IF $Z_f Z_i$ ARE RESISTIVE, THE AMPLIFIER IS WIDEBAND WITHIN THE LIMITATION OF THE OPERATIONAL AMPLIFIERS GAIN BANDWIDTH PRODUCT.

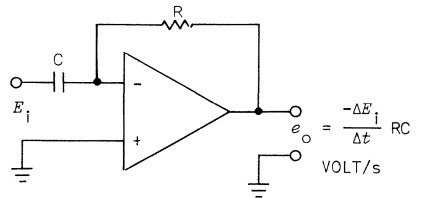
IF $Z_f Z_i$ ARE RC COMBINATIONS THE AMPLIFIER FREQUENCY RESPONSE WILL DEPEND ON THE RC RATIO.

IF $Z_f Z_i$ ARE NONLINEAR WITH VOLTAGE OR CURRENT (DIODE) THEN THE AMPLIFIER IS NONLINEAR, EITHER COMPRESSION OR EXPANSION.



OUTPUT PROPORTIONAL TO INPUT VOLTAGE AND TIME

INTEGRATOR



OUTPUT PROPORTIONAL TO RATE OF CHANGE OF INPUT VOLTAGE

DIFFERENTIATOR

Fig. 20-9 Operational amplifier applications using the inverting (-) input to the amplifier.

current to
voltage
converter

The previous circuits use the noninverting + input to an operational amplifier as well as the inverting - input. The following circuits use only the - input to the operational amplifier with the + input grounded. The basic operational amplifier and feedback impedance discussed in Section 20.6 is essentially a current to voltage converter as shown in Fig. 20-9 which provides an output voltage proportional to the input current and feedback impedance.

inverting
amplifier

The input current for a current to voltage converter circuit may be obtained from an input voltage by the use of an input impedance resulting in the inverting amplifier shown in Fig. 20-9. The output voltage is then proportional to the input voltage and to the ratio between the feedback impedance and input impedance. By selecting appropriate impedance values this inverting amplifier may also be used as an inverting attenuator.

In most applications the input impedance and feedback impedance used for an inverting amplifier are resistors and the amplifier is essentially wideband.

For large $\frac{Z_f}{Z_i}$ ratios the gain of the circuit will be

compensating
adapter

high and the use of resistors for Z_i and Z_f may produce some high frequency losses due to stray capacitance. These losses can be minimized by the use of the Tektronix Compensating Adapter (Part Number 013-0081-00). This adapter compensates for the stray capacitance associated with the internal Z_i and Z_f resistors.

bandwidth
modification

If the Z_i and Z_f components of an inverting amplifier are combinations of resistors and capacitors, then these impedances will have nonlinear frequency response characteristics and the amplifier may therefore be tailored to suit particular frequency response requirements. Various values of resistors and capacitors for the input impedance and feedback impedances can be chosen to provide high-pass filtering, low-pass filtering or other bandwidth modification in a biophysical measurement system.

logarithmic
display

If the input impedance and feedback impedance elements in an inverting amplifier are nonlinear with respect to voltage or current then the amplifier's gain will be nonlinear with respect to voltage or current. If the Type 3A8 is used with a Tektronix Log Adapter (PN 013-0067-00) the relationship between the input voltage and the output voltage of the operational amplifier will be logarithmic and thus allows the measurement of high amplitude signals mixed with low amplitude signals. Thus, biophysical signals spanning three voltage decades can be displayed logarithmically on an oscilloscope.

20.8 INTEGRATION AND DIFFERENTIATION WITH OPERATIONAL AMPLIFIERS

Operational amplifier circuits are particularly suited to integration and differentiation applications, that is, where the output voltage is proportional to either the integral or the differential of the input voltage.

integration

An operational amplifier used as an integrator is shown in Fig. 20-9. The output voltage is proportional to the integral of the input voltage in volt-seconds during the time interval t_1-t_2 . In integration it is necessary to specify this time interval, for the output voltage will continue to increase as long as a positive input voltage is present and, when the input voltage is removed, the output voltage will remain at a fixed level. In order to return this integrator output to zero, allowing the integrator to once more perform useful integration, the feedback capacitor of the integrator must be discharged either by a manual switch, a reed switch controlled by other circuitry or by the use of the Tektronix Gating Adapter (PN 013-0068-00). This gating adapter is designed for use with one of the operational amplifiers in the Type 3A8 and allows repetitive signals with a net integral other than zero to be integrated and displayed. The use of an integrator in biophysical measurements is further discussed in Chapter 12, Section 12.6, when discussing the electromyogram produced by a voluntary muscular action. Details of a simple gating adapter for use with an integrator are given in Chapter 29.

differenti-
ation

An operational amplifier may be used for differentiation as shown in Fig. 20-8B. The output voltage is then proportional to the differential of the input voltage, i.e., it is proportional to the rate of change of the input voltage in volts per second. Differentiation is rarely used directly in biophysical measurement systems due to the excessive noise it produces, however, indirectly it may be used to intensity compensate an oscilloscope display as discussed in Chapter 21.

The values of R and C used in integration and differentiation circuits depend upon the amplitude and frequency of the input signal and are usually selected experimentally to produce an adequate output voltage without attempting to exceed the operational amplifier's ± 25 -V output capability.

OSCILLOSCOPES

A physiological signal, after having been amplified and processed, can be displayed and/or recorded on many different devices as shown in Chapter 15, Fig. 15-1. The oscilloscope is the most popular of these devices and is incorporated within almost every biophysical instrumentation system. In many cases, the oscilloscope is the final component in an instrumentation system. However, as an oscilloscope does not provide a permanent record, some form of recorder, such as a magnetic tape recorder, a graphic recorder or an oscilloscope camera, may also be necessary.

This chapter should not be regarded as a general reference on oscilloscopes. It specifically discusses the unique characteristics of oscilloscopes as they relate to biophysical measurements. Many companion volumes relating to either oscilloscope measurement concepts or oscilloscope circuit concepts are available from Tektronix; a complete list of these volumes is given at the end of this book. We would also recommend *Typical Oscilloscope Circuitry*, published by Tektronix, as a good single reference on oscilloscope principles. Also, the Tektronix catalog of oscilloscopes and associated instruments provides a useful reference while listing the characteristics of all of Tektronix' products. This catalog includes a list of US field engineering offices and international field offices and distributors where additional information on oscilloscopes and their applications are available. Tektronix also produces an applications oriented periodical entitled *TEKSCOPE*. For regular receipt of this periodical, contact the before-mentioned field office or distributor in your area.

reference
material

The three principal components of a simple oscilloscope are the vertical amplifier, the horizontal amplifier/sweep generator and the cathode-ray tube. These components are discussed in detail in the before-mentioned references and the characteristics of amplifiers suited for use with physiological signals are further discussed in Chapter 19. An oscilloscope may contain an amplifier suitable for displaying a physiological signal directly from electrodes or transducers (such as the Type 5031) or it may incorporate a simpler amplifier which can be used in conjunction with a preamplifier for physiological applications.

classifi- cations	It is important to differentiate between three essentially similar devices that present information on a cathode-ray tube: the conventional oscilloscope, the slave oscilloscope and the display device. A conventional oscilloscope is a versatile instrument incorporating a vertical amplifier and a horizontal amplifier/sweep generator in conjunction with a CRT display. Both the vertical amplifier and the sweep generator's characteristics can be varied over wide ranges by controls on the front of the oscilloscope. A slave oscilloscope's function is to provide a conventional oscilloscope with an additional CRT display. The slave oscilloscope is coupled to a conventional oscilloscope in such a way as to present the same display on its CRT as presented on the CRT of the conventional oscilloscope. The coupling between a slave oscilloscope and "master" conventional oscilloscope is usually achieved with high impedance probes limiting the physical distance between two oscilloscopes to 12 feet. A display device is somewhat similar to a slave oscilloscope; however, it is intended to be used in a remote location for the display of any signals on x - y - z coordinates. These signals may be derived via low impedance outputs from an oscilloscope or they may be derived from other instrumentation systems.
conventional scope	
slave scope	
display device	

21.1 OSCILLOSCOPE VERTICAL AMPLIFIERS

To preserve linearity and focus on a CRT display, it is necessary to drive both the CRT vertical deflection plates and horizontal deflection plates with differential signals. The oscilloscope amplifier in its simplest form may simply amplify an input signal and present this amplified signal to the CRT deflection plates as a differential signal.

To obtain various deflection sensitivities the gain of this amplifier must be variable both in discrete steps and over a continuous range. A simple oscilloscope vertical amplifier, such as the Tektronix Type 3A75 Amplifier, accepts an input signal into a 1-M Ω impedance and attenuates this signal by discrete steps to 50 mV/cm of deflection on the CRT display. The 3A75 then amplifies the signal and provides a differential output of +10 V/cm and -10 V/cm to the CRT deflection plates. The CRT thus receives a total deflection voltage of 20 V/cm.

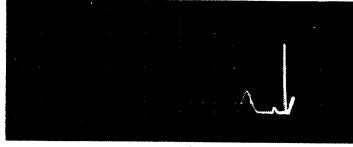
multitrace
scopes

Multitrace amplifiers allow the presentation of several channels of information by switching a CRT display between several amplifiers in such a manner that the display appears to a viewer as being produced by several separate CRT deflection systems within the one CRT. The Tektronix Type 3A3 Dual Trace Differential Amplifier switches or "chops" between two amplifiers at a 0.2 MHz rate. The CRT display thus consists of a sequence of 2.5 μ s segments from each channel in turn. At the maximum sweep speed commonly used for physiological signals of 0.1 ms/cm, the two traces on the CRT appear as continuous lines since 20 chopped segments occur during each centimeter of sweep and therefore the separate segments cannot be discerned. Four-trace amplifiers such as the Type 3A74 incorporate switching circuitry to switch between four separate amplifiers, presenting a four-channel display on a CRT.

21.2 OSCILLOSCOPE HORIZONTAL AMPLIFIERS/SWEEP GENERATORS

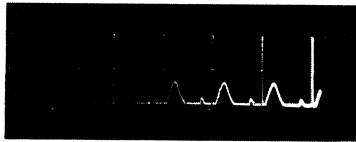
The oscilloscope horizontal system, as with the vertical system, requires a differential voltage to deflect the CRT beam and is similar to the vertical amplifier with the exception that the variable and step gain controls are often omitted (a constant level of sweep signal is usually used in conjunction with the horizontal amplifier). A sweep generator provides a ramp to the horizontal amplifier, thus causing a linear horizontal sweep across the CRT face. For physiological applications the sweep generator ramp duration should be variable from 50 seconds to less than 1 ms, providing sweep speeds from 5 s/div to 0.1 ms/div. Some applications may require faster sweep speeds; most Tektronix sweep generators are capable of sweeping to at least 1 μ s/div.

A SIMULATED ECG DISPLAYED AT 0.5s/DIV



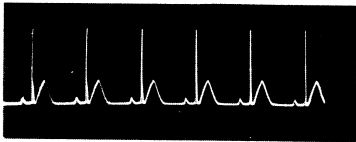
USABLE PERSISTENCE FOR 1.5s MAXIMUM

(A) OSCILLOSCOPE WITH P31 PHOSPHOR



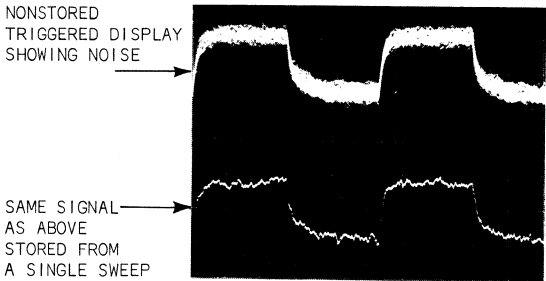
USABLE PERSISTENCE FOR 4s MAXIMUM

(B) OSCILLOSCOPE WITH P7 PHOSPHOR



EFFECTIVELY HAS INFINITE PERSISTENCE

(C) STORAGE OSCILLOSCOPE SUCH AS
TYPE 564B



(D) STORAGE DISPLAYS MAY HELP TO REDUCE APPARENT NOISE ON A TRACE AS SHOWN BY THE NONSTORED AND STORED DISPLAYS PRESENTED ABOVE

Fig. 21-1. Persistence and storage.

21.3 OSCILLOSCOPE CRT DISPLAYS

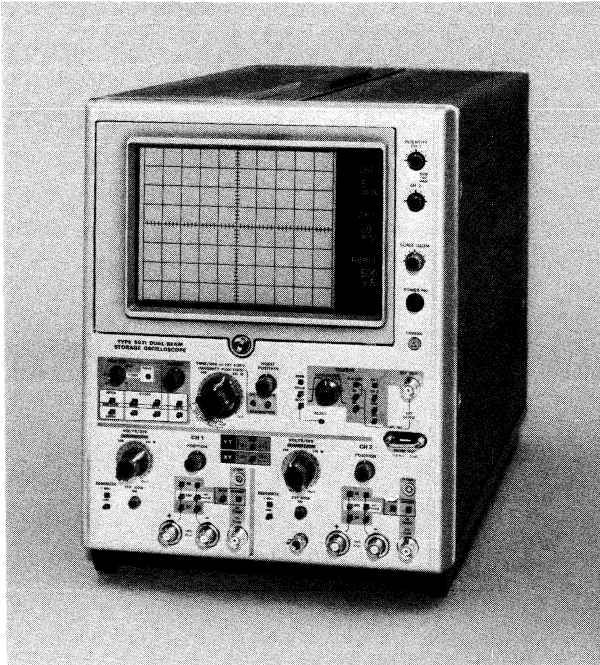
Most conventional oscilloscope CRT displays are intended to be viewed at a distance from 2 feet to 6 feet and are presented in a format approximately 10 centimeters wide and 8 centimeters high. Special large screen CRT displays are available for viewing at greater distances or for viewing by a number of observers.

persistence The persistence of a CRT display refers to the length of time that the CRT phosphor continues to emit light after the CRT beam has excited the phosphor. Referring to Fig. 21-1A, the standard phosphor supplied with many Tektronix oscilloscopes is a P31 phosphor which provides usable persistence for up to 1.5 seconds; however, for most practical purposes its persistence is considered to be under 0.5 seconds. Most Tektronix oscilloscopes are available on special order with a Type P7 phosphor (see Fig. 21-1B) which provides considerably greater persistence than the P31 phosphor.

storage A storage oscilloscope such as the Tektronix Type 564B or Type 5031 effectively provides infinite persistence (see Fig. 21-1C) and should be considered for all oscilloscope applications requiring sweep speeds below 10 ms/div or where the information displayed is transient in nature. The storage oscilloscope stores a trace on the CRT phosphor until such time as an erase control is manually activated or activated by a remote electrical signal. As shown in Fig. 21-1D, the storage oscilloscope may also be useful in place of a conventional oscilloscope in reducing the apparent noise on a CRT display.

21.4 TYPICAL OSCILLOSCOPES

- Type 5031 Oscilloscopes representative of Tektronix products suited to biophysical measurements are shown in Fig. 21-2, 21-3 and 21-4. The Tektronix Type 5031 Dual Beam Storage Oscilloscope shown in Fig. 21-2 is a non-plug-in storage oscilloscope providing a general purpose instrument particularly suited to biophysical measurements. This oscilloscope is referred to as a dual-beam oscilloscope, having two separate vertical deflection systems and two separate pairs of vertical deflection plates within the CRT. This contrasts to the dual-trace amplifier discussed earlier that presents two channels of information via a beam switching circuit. Each vertical amplifier has a maximum sensitivity of $10 \mu\text{V}/\text{div}$ with a differential input impedance of $2 \text{ M}\Omega$ and a common mode input impedance of $0.5 \text{ M}\Omega$ (modifiable to $5 \text{ M}\Omega$). No electrodes or other input coupling items are provided as standard accessories with this product.
- 560 series The Tektronix 560 Oscilloscope series shown in Fig. 21-3 is representative of the plug-in oscilloscope concept. Both the vertical amplifier and the horizontal amplifier/sweep generator characteristics in the 564B Storage Oscilloscope and 561B conventional Oscilloscope can be varied over broad ranges by the use of various vertical and time-base plug-in units. Fig. 21-3 shows only those plug-in units that are suited to biophysical measurements. The 560 series of oscilloscopes includes many more vertical and time-base plug-in units for specialized, non-physiological applications.
- The Type 565 Oscilloscope shown in Fig. 21-3 does not utilize a time-base plug-in unit; however, it is a dual-beam oscilloscope (having two separate and independent deflection systems) utilizing two vertical plug-in units. A Type 565, when used in conjunction with, for example, two Type 3A74 four-trace plug-in units, can present a total of eight traces on its CRT.



ALSO AVAILABLE IN A RACK
MOUNTING OR LOW PROFILE
CABINET CONFIGURATION.

TYPE R5031

ALSO AVAILABLE AS A
NONSTORAGE OSCILLOSCOPE
IN A RACK MOUNTING OR
LOW PROFILE CABINET
CONFIGURATION

TYPE R5030

BISTABLE SPLIT-SCREEN STORAGE AND CONVENTIONAL DISPLAYS
WITH VARIABLE VIEWING TIME AND AUTOMATIC ERASE

SCALE FACTOR READOUT ADJACENT TO THE CRT

DUAL BEAM DIFFERENTIAL. EACH BEAM TO $10\mu\text{V}/\text{DIV}$ AT 1MHz
OR 5kHz WITH 100,000:1 CMRR

SELECTABLE X-Y MODE AT $10\mu\text{V}/\text{DIV}$

Fig. 21-2. The Tektronix Type 5031 Dual Beam Storage Oscilloscope.

VERTICAL PLUG-IN UNITS

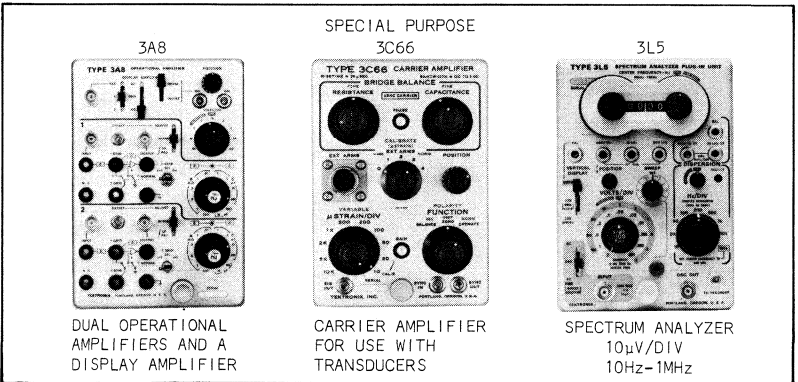
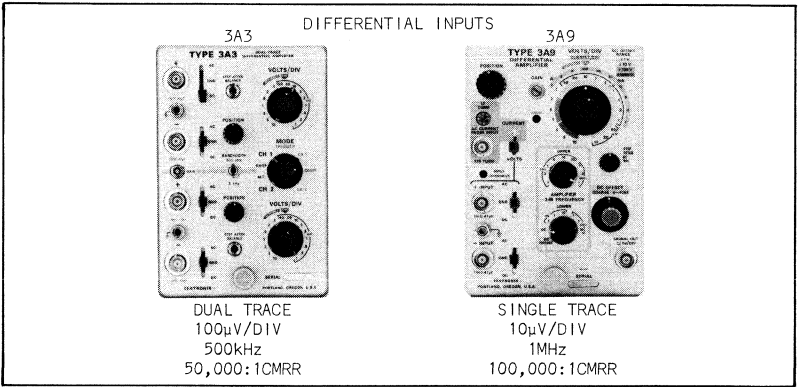
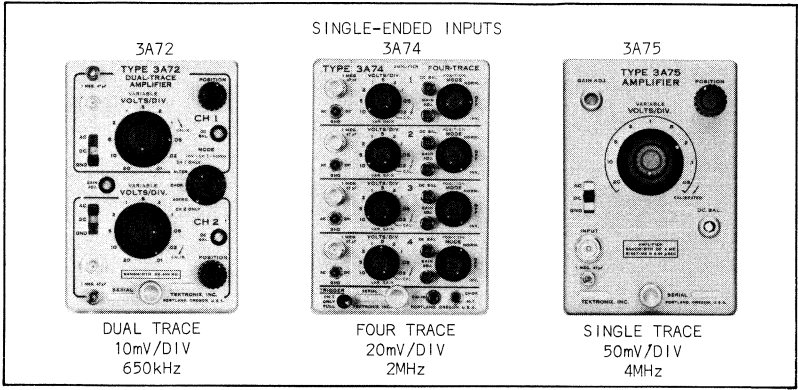
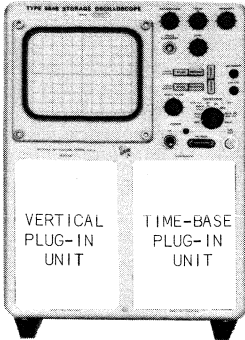


Fig. 21-3. Part of the Tektronix 560 Series of oscilloscopes.

PLUG-IN OSCILLOSCOPES



564B

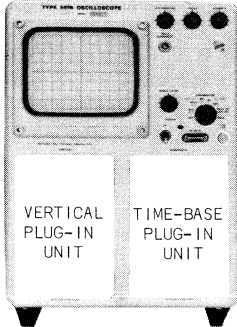
BISTABLE SPLIT-SCREEN STORAGE AND CONVENTIONAL DISPLAYS

AVAILABLE WITH VARIABLE VIEWING TIME AND AUTO ERASE

USES INTERCHANGEABLE VERTICAL AND TIME-BASE PLUG-IN UNITS

VERTICAL
PLUG-IN
UNIT

TIME-BASE
PLUG-IN
UNIT



561B

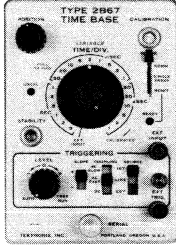
CONVENTIONAL NONSTORAGE DISPLAYS ONLY

USES INTERCHANGEABLE VERTICAL AND TIME-BASE PLUG-IN UNITS

VERTICAL
PLUG-IN
UNIT

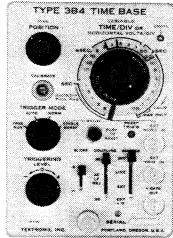
TIME-BASE
PLUG-IN
UNIT

TIME-BASE PLUG-IN UNITS



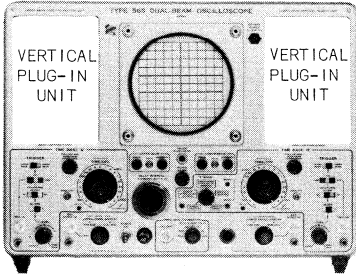
2B67

1 μ s/DIV-5s/DIV
5X MAGNIFIER
SINGLE SWEEP
LIMITED USE AS AN AMPLIFIER



3B4

0.2 μ s/DIV-5s/DIV
50X MAGNIFIER
SINGLE SWEEP
AS AN AMPLIFIER
0.2V/DIV-5V/DIV



565

DUAL-BEAM OSCILLOSCOPE HAVING TWO VERTICAL AND TWO HORIZONTAL SYSTEMS. CONVENTIONAL NONSTORAGE DISPLAY ONLY

USES INTERCHANGEABLE VERTICAL PLUG-IN UNITS

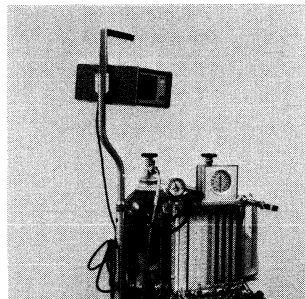
HORIZONTAL SYSTEMS - 1 μ s/DIV-5s/DIV
10X MAGNIFIER
SWEEP DELAY
LIMITED USE AS AN AMPLIFIER

VERTICAL
PLUG-IN
UNIT

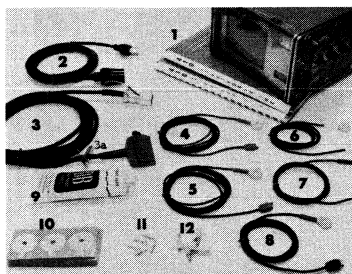
VERTICAL
PLUG-IN
UNIT



MONITORS ECG, EEG OR RELATIVE BLOOD PRESSURE
ALARM ON LOSS OF ECG OR PRESSURE SIGNAL
BATTERY OPERATED PORTABLE - 12.5 POUNDS

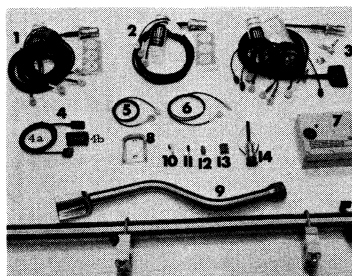


THE TYPE 410 MONITOR POSITIONED AT THE 5ft
LEVEL ON AN ANESTHESIOLOGIST'S GAS
MACHINE DEMONSTRATING USE OF THE MOUNTING
STAND SHOWN AS AN OPTIONAL ACCESSORY



THE TYPE 410 WITH STANDARD ACCESSORIES

REF	QTY	DESCRIPTION	PART NUMBER
1	2	INSTRUCTION MANUALS	070-0658-00
2	1	POWER CABLE ASSEMBLY - 7ft	161-0058-00
3	1	PATIENT CABLE ASSEMBLY - 10ft	012-0120-00
3A		INCLUDES SHEET CLAMP	344-0146-00
4	1	ELECTRODE RL GREEN - 6ft	012-0121-25
5	1	" LL RED - 6ft	012-0121-22
6	1	" RA WHITE - 4ft	012-0121-29
7	1	" LA BLACK - 4ft	012-0121-20
8	1	" C BROWN - 4ft	012-0121-21
9	1	TUBE ELECTRODE PASTE	006-1098-00
10	1	PACKAGE 100 ADHESIVE RINGS	006-1099-00
11	5	PLATE ADAPTER	103-0079-00
12	5	NEEDLE ADAPTERS	103-0108-00



OPTIONAL ACCESSORIES FOR USE WITH THE
TYPE 410 MONITOR

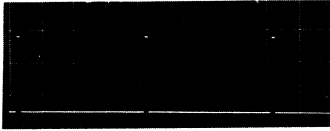
REF	DESCRIPTION	PART NUMBER
1	LIMB LEAD CABLE PACKAGE LIMB LEAD CABLE WITH FOUR ELECTRODES INSTALLED, ELECTRODE PASTE AND ADHESIVE RINGS (FOR ECG MONITORING IN SURGERY)	012-0184-00
2	CHEST LEAD CABLE PACKAGE CHEST LEAD CABLE WITH THREE ELECTRODES INSTALLED, ELECTRODE PASTE AND ADHESIVE RINGS (FOR ECG MONITORING IN INTENSIVE CARE)	012-0183-00
3	PATIENT CABLE PACKAGE SAME AS STANDARD ACCESSORY PACKAGE (ABOVE) LESS POWER CORD AND MANUALS.	012-0185-00
4	PULSE SENSOR ASSEMBLY	015-0104-00
4A	CONSISTING OF PULSE SENSOR	015-0102-00
4B	AND FINGER HOLDER	015-0103-00
5	ELECTRODE +EEG YELLOW -2½ft	012-0121-24
6	" -EEG " 2½ft	012-0121-23
7	BATTERY PACK	016-0107-02
8	MOUNTING CUP	407-0393-01
9	MOUNTING STAND	016-0110-00
10	BARE WIRE ADAPTER	103-0080-00
11	PLUG - MINI-PHONE	134-0079-00
12	PLUG - 2 P IN ELECTRODE	134-0089-00
13	PLUG - 7 P IN AUXILIARY	134-0090-00
14	PLUG OUTPUT	131-0551-01

Fig. 21-4. The Tektronix Type 410 Physiological Monitor.

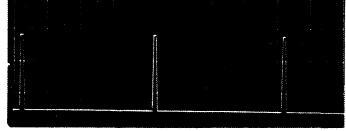
Type 410

The Tektronix Type 410 Physiological Monitor shown in Fig. 21-4 is a specialized oscilloscope intended for monitoring during surgery or in intensive care. The Type 410 provides a minimum of operator controls to allow its use by nontechnical personnel within the hospital environment. A full complement of electrodes and a patient cable are provided as standard accessories with many optional accessories available as shown in Fig. 21-4.

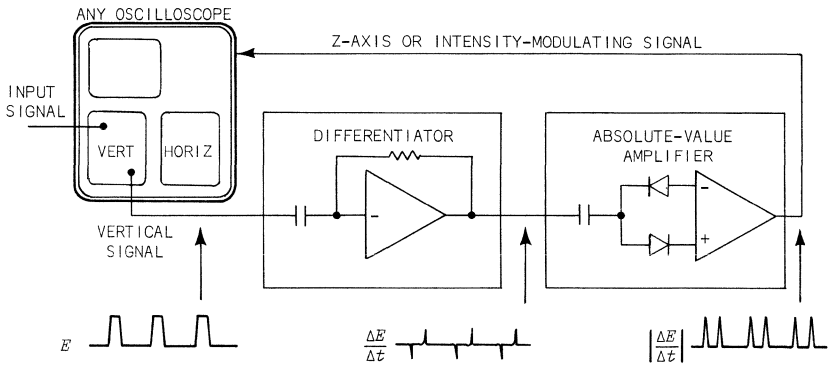
Full specifications on the oscilloscopes discussed here are given in the current Tektronix catalog of oscilloscopes and associated instruments.



A STANDARD OSCILLOSCOPE DISPLAY OF A 0.5ms WIDE PULSE AT A SWEEP SPEED OF 5ms/cm (NO RATE INTENSIFICATION)

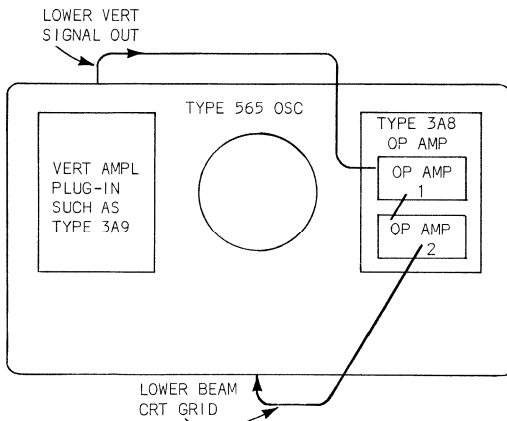


OSCILLOSCOPE DISPLAY OF THE SAME WAVEFORM AS SHOWN AT LEFT WITH RATE INTENSIFICATION



IF NEGATIVE INTENSITY MODULATING SIGNAL IS REQUIRED, REVERSE DIODES IN ABSOLUTE-VALUE AMPLIFIER

(A) PRINCIPLE



OPERATIONAL AMPLIFIER 1 OPERATES AS DIFFERENTIATOR WITH $Z_1 = .001\mu f$ AND $Z_f = 200k\Omega$

OPERATIONAL AMPLIFIER 2 OPERATES AS AN ABSOLUTE-VALUE AMPLIFIER USING THE ABSOLUTE-VALUE ADAPTER DESCRIBED IN CHAPTER 29

(B) A TYPICAL RATE INTENSIFIED SYSTEM

Fig. 21-5. Rate intensification.

21.5 RATE INTENSIFICATION

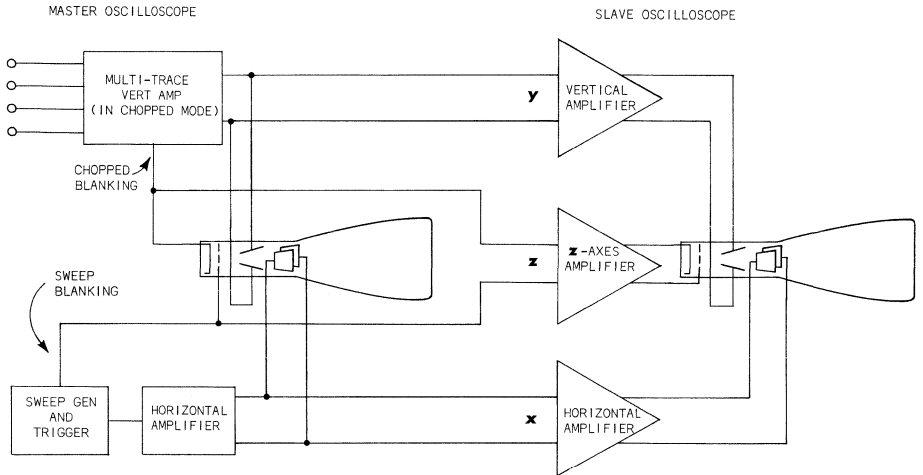
The left-hand oscilloscope waveform shown in Fig. 21-5 is typical of a pulse display on an oscilloscope. Only the baseline and top of the pulse are displayed on the CRT because during the transitions, the CRT beam is moving 100 times faster than during the baseline or pulse top periods. This difference in rate results in a difference in intensity on the CRT display.

differenti-
ation and
absolute
value
amplification

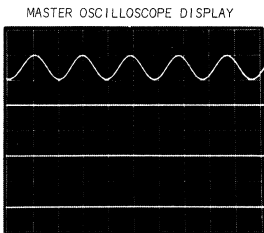
In order for the baseline, the pulse top and the transitions to be all presented on the CRT at the same intensity, the intensity of the CRT display must be made proportional to the rate of change of the vertical information. Fig. 21-5A shows the principle of rate intensification. A differentiator provides a signal proportional to the rate of change of the vertical signal. The output on the differentiator is, however, direction sensitive and must be "full-wave rectified" in an absolute-value amplifier to produce an output proportional to the absolute rate of change of the vertical signal. This output is then used in conjunction with the z-axis input of an oscilloscope to intensity modulate the CRT beam.

practical
system

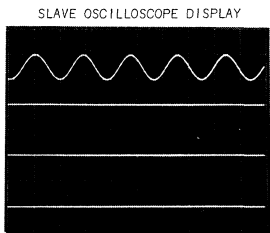
The above principle can be applied in practice as shown in Fig. 21-5B. The CRT photographs shown in Fig. 21-5 were taken with this system. The differentiator used consists of one of the operational amplifiers in a Type 3A8 Operational Amplifier plug-in unit operating as a differentiator. The differentiator's input impedance and feedback impedance are selected by the controls on the front of the 3A8 at $.001 \mu\text{F}$ and $0.2 \text{ M}\Omega$, respectively. The output from this operational amplifier is then fed to the second operational amplifier on the Type 3A8 which is used as an absolute-value amplifier in conjunction with the absolute-value adapter described in Chapter 29. The output from this absolute-value amplifier is then fed into the lower-beam CRT grid to modulate the lower beam intensity. Rate intensification is particularly useful when attempting to photograph a CRT display or when many channels of information are presented on a single CRT display.



- y** CHANNEL BANDWIDTH MUST BE SUFFICIENT TO DISPLAY CHOPPING INFORMATION WITH NEGLIGIBLE DISTORTION
- z** CHANNEL BANDWIDTH AND DELAY MUST BE ADEQUATE TO PRODUCE EFFECTIVE CHOPPED BLANKING
- x** IF MAXIMUM SWEEP SPEED IS $< 0.1\text{ms/cm}$ BANDWIDTH OF 100kHz IS ADEQUATE



FOUR CHANNELS DISPLAYED IN CHOPPED MODE WITH SINEWAVE INPUT TO CHANNEL 1



0.1ms/cm

Fig. 21-6. Oscilloscope slaving considerations.

21.6 SLAVE OSCILLOSCOPES

additional
CRT display

The purpose of the slave oscilloscope, as mentioned previously, is to provide a conventional oscilloscope with an additional CRT display. A slave oscilloscope should, thus, contain only those controls necessary to alter the characteristics of the slave oscilloscope display, that is, intensity, focus and perhaps astigmatism and scale illumination. Although a slave oscilloscope will contain a vertical amplifier, a horizontal amplifier and a z -axis amplifier, it is unnecessary to vary the characteristics of these amplifiers after the slave oscilloscope has been coupled to the master oscilloscope; thus, these amplifiers need no external controls. A block diagram of a slave oscilloscope system is shown in Fig. 21-6 and the ideal relationship between a master oscilloscope display and the slave oscilloscope display is also shown in Fig. 21-6. Any change in the master oscilloscope display should be perfectly reproduced by the slave oscilloscope display.

nondistorting
coupling

high cost

When considering slave oscilloscopes, two basic rules should generally be adhered to. First, the required signal from the master oscilloscope should be coupled to the slave oscilloscope via high-impedance, low-capacity probes so as not to interfere with the characteristics of the master oscilloscope. Tektronix X10 or X100 probes are recommended as, even with 12-foot lengths, they provide a load on the master oscilloscope of 10 M Ω and less than 13 pF. Invariably, the master oscilloscope is not designed to allow low-impedance probes or direct interconnection and the use of either will degrade the performance of the master oscilloscope. The second basic rule of slave oscilloscopes relates to economy. In general, the cost of a master oscilloscope/slave oscilloscope system will be equal to or will exceed the cost of two separate oscilloscopes. Thus, the criteria for using a slave oscilloscope system should not be based on economy but rather on the desirability of having one set of operator controls controlling two CRT displays.

vertical and
horizontal
slaving

The slave oscilloscope block diagram shown in Fig. 21-6 shows both the vertical (y) and horizontal (x) signals coupled differentially from the master oscilloscope deflection plates to the slave oscilloscope deflection amplifiers. It is desirable to use differential inputs to eliminate unwanted signals that may appear on the master oscilloscope amplifier outputs but may not appear on the master oscilloscope display due to the inherent differential rejection capability of the deflection plates. If a multitrace vertical amplifier is used in the master oscilloscope vertical, the slave oscilloscope vertical amplifier must have sufficient bandpass to allow the chopped vertical signal to be coupled into the slave oscilloscope without risetime reduction. For most Tektronix multitrace vertical amplifiers, this necessitates a slave oscilloscope vertical amplifier bandpass considerably in excess of 1 MHz. The bandwidth of the slave oscilloscope horizontal amplifier is noncritical in biophysical measurements applications and 0.1 MHz is usually sufficient.

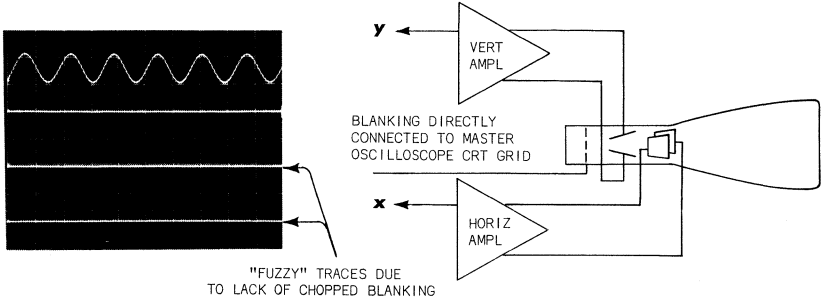
sweep and
chopped
blanking
slaving

A master oscilloscope display uses intensity modulation via the CRT grid and/or cathode to turn the master oscilloscope CRT beam off during sweep retrace and during the interval that the multitrace vertical amplifier is switching from one channel to another. These signals are referred to as sweep blanking and chopped blanking, respectively. The z -axis amplifier must be capable of accepting both of these blanking signals to allow the slave oscilloscope CRT beam to be turned off in synchronism with the master oscilloscope CRT beam. The chopped blanking signal consists of a narrow pulse occurring during the chopping interval; this narrow pulse has a risetime considerably less than 1 μ s, thus, a z -axis amplifier bandwidth of several megahertz is necessary to allow faithful chopped

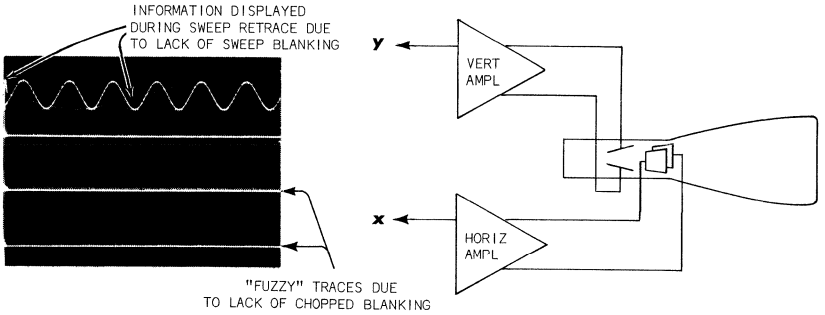
blanking of the slave oscilloscope display. If a single trace vertical amplifier is used, then no chopped blanking signal is involved and only the sweep blanking need be coupled to the z -axis amplifier. Under these conditions the bandwidth of the z -axis amplifier would be noncritical for biophysical measurement applications and, as with the horizontal amplifier, 0.1 MHz should be sufficient.

Various compromises may be made to the basic slave oscilloscope block diagram shown in Fig. 21-6; however, it is important to fully understand any performance degradation associated with these compromises. All Tektronix oscilloscopes utilize regulated power supplies and high quality vertical and horizontal amplifiers. Thus, it is not absolutely necessary to couple the vertical and horizontal signals from the master oscilloscope to the slave oscilloscope in a differential mode. Negligible distortion will be observed on the slave oscilloscope display if the signals are coupled between master and slave with single probes. It should be noted, however, that the output from the master oscilloscope vertical and horizontal amplifiers will contain a standing DC level and it is necessary for the slave vertical and horizontal amplifiers to be able to cancel out this standing DC level if only single-probe coupling is used. Most Tektronix vertical and horizontal amplifiers suitable for use with slave oscilloscopes incorporate sufficient range in the DC BALANCE control and POSITION control to cancel out this DC voltage. It is, however, necessary to fully consider this and to ensure that sufficient position range is available in the slave oscilloscope to center the display on the slave CRT.

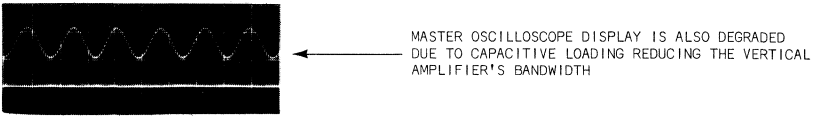
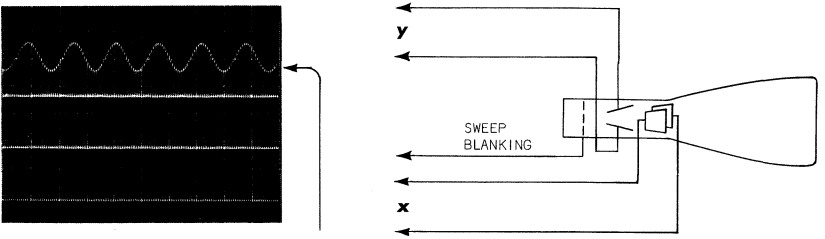
differential
versus
single-probe
coupling



(A) DIRECT SWEEP-BLANKING INTERCONNECTION. SINGLE PROBES USED TO PICK OFF **x** AND **y** INFORMATION.



(B) NO **x** AXIS INFORMATION. SINGLE PROBES USED TO PICK OFF **x** AND **y** INFORMATION



(C) DIRECT INTERCONNECTION OF **x**, **y** AND SWEEP BLANKING BETWEEN MASTER AND SLAVE - 12ft LONG UNSHIELDED WIRES.

Fig. 21-7. Oscilloscope slaving compromises.

fuzziness

Many slave oscilloscopes are used with multitrace vertical amplifiers in the master oscilloscope but with no chopped blanking signal coupled between the master oscilloscope and the slave oscilloscope. No chopped blanking is, therefore, presented to the slave CRT resulting in a fuzziness on the traces as shown in Fig. 21-7A and B. Due to the characteristics of the film used to record the waveform shown in Fig. 21-7A and B, this fuzziness does not appear to be excessive; however, a viewer would find the lack of chopped blanking quite annoying when viewing the slave oscilloscope display.

no sweep blanking

If a slave oscilloscope system does not incorporate sweep blanking, as shown in Fig. 21-7B, sweep retrace can clearly be observed on the slave oscilloscope CRT and under certain input signal conditions can be misleading as well as annoying.

paralleling

Many attempts have been made to create a slave oscilloscope by simply paralleling an additional CRT with the master oscilloscope CRT; the results, as shown in Fig. 21-7C, are far from ideal. This system also degrades the performance of the master oscilloscope as well as presenting a poor display on the slave oscilloscope and is, therefore, not generally recommended. Also, as the additional CRT will not have exactly the same characteristics as the master oscilloscope CRT, the slave oscilloscope display will never be a direct replica of the master oscilloscope display.

slave
systems

Two practical slave oscilloscope systems are shown in Figs. 21-8 and 21-9. The single-channel slave oscilloscope system shown in Fig. 21-8 uses either a Type 601 Storage Display Unit or a Type 602 (conventional) Display Unit as a slave oscilloscope. The master oscilloscope consists of either a Type 561B (Conventional) Oscilloscope or a Type 564B Storage Oscilloscope with a 2B67 Time Base plug-in unit and any single-channel vertical amplifier plug-in unit. The x and y bandwidth of the slave oscilloscope is 1 MHz, sufficient for a single-channel operation. The y input to the slave oscilloscope is derived via a single probe from the output of the vertical amplifier. The position control within the slave oscilloscope is sufficient to cancel out the standing DC level at the output of the vertical amplifier. The x input to the slave oscilloscope is derived in the same manner as for the y input. As the system is only intended for single-channel operation, no chopped blanking is required and the z input to the slave oscilloscope must simply provide sweep retrace blanking. The z input of the slave oscilloscope is derived via a probe from the blanking pulse output of the time base plug-in unit. In order to provide sufficient gain range in the 601 or 602 display unit, R25 and R75 in these units must be shorted out with a wire link.

probe
coupling

The Type 601 and 602 Storage Display Units have 100,000-ohm input impedance amplifiers. Tektronix P6006 (X10) and P6007 (X100) probes are intended for use with amplifiers having a 1-megohm input impedance. These probes can, however, be successfully used with the 601 or 602 display unit; the X10 probe actually providing an attenuation of X100 and the X100 probe actually providing an attenuation of X200. It should be noted that the compensation on all probes used in any slave oscilloscope system must be correctly adjusted. Correct probe compensation adjustment for probes used with the z -axis amplifier may require an additional oscilloscope to view the z -axis waveform during adjustment.

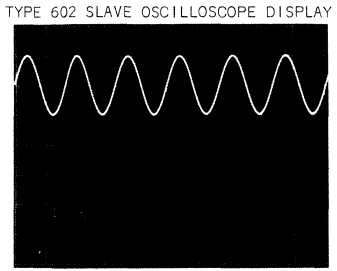
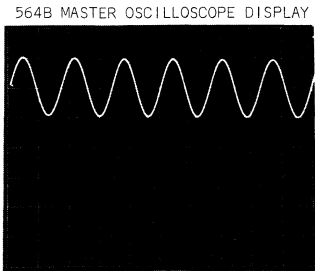
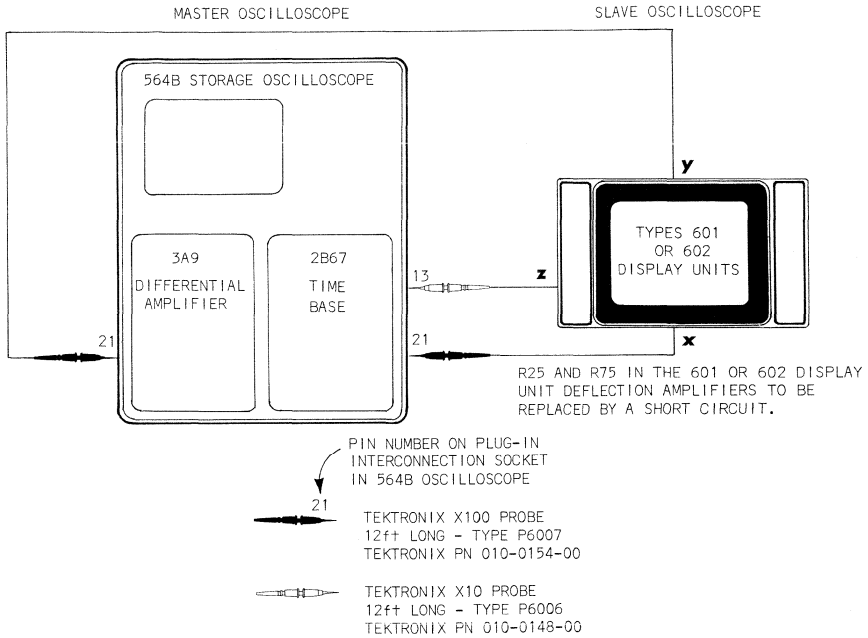
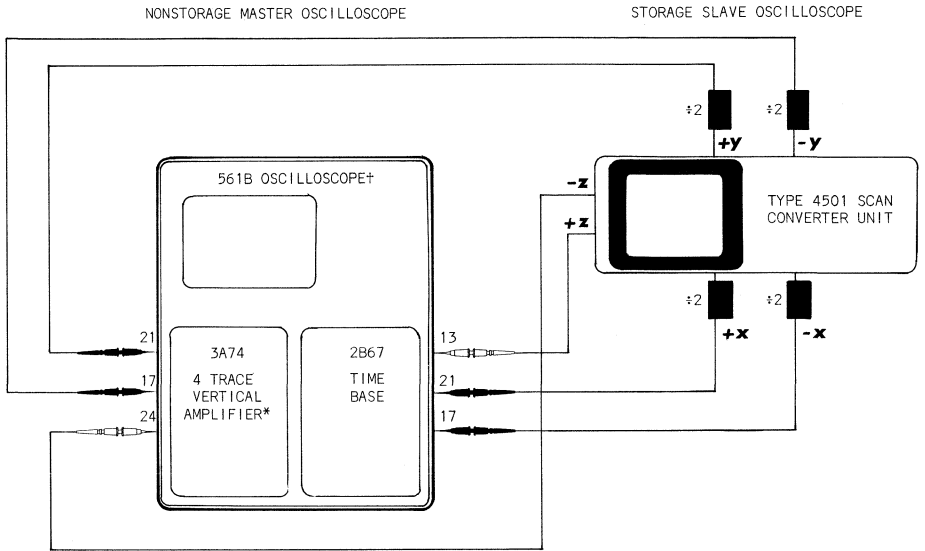





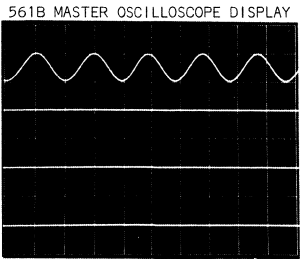
Fig. 21-8. A single channel slave oscilloscope system.



*ANY PLUG-IN TYPES DESIGNED FOR USE WITH TEKTRONIX 560 SERIES OSCILLOSCOPES MAY BE USED

+MAY ALSO USE TYPE 564B STORAGE OSCILLOSCOPE

- PIN NUMBER ON PLUG-IN INTERCONNECTION SOCKET IN 561B OSCILLOSCOPE
-  21 TEKTRONIX X100 PROBE
12ft+ LONG - TYPE P6007
TEKTRONIX PN 010-0154-00
 -  17 TEKTRONIX X10 PROBE
12ft+ LONG - TYPE P6006
TEKTRONIX PN 010-0148-00
 -  #2 TEKTRONIX #2 ATTENUATOR
"INPUT NORMALIZER"
TEKTRONIX PN 067-0541-00



NOTE: THE TYPE 4501 SCAN CONVERTER PROVIDES EITHER A NON-STORAGE OR A STORAGE DISPLAY.

WHEN USING THE TYPE 4501 TO PROVIDE A STORED DISPLAY, THE 4501 PROVIDES THIS DISPLAY IN VIDEO FORMAT FOR DISPLAY ON A TELEVISION RECEIVER OR MONITOR. REFER TO CHAPTER 23.

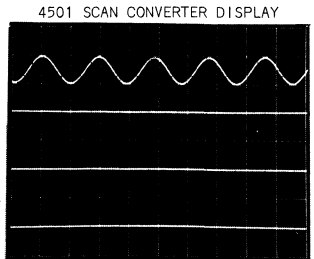


Fig. 21-9. A multichannel slave oscilloscope system.

Type 4501
slave
system

A multichannel slave oscilloscope system is shown in Fig. 21-9. The Type 4501 Scan Converter Unit is basically a five-inch storage display unit having high input impedance ($1\text{ M}\Omega$, 47 pF) differential input into the x , y , and z axes. The bandwidth of the x and y amplifiers is 10 MHz and the bandwidth of the z -axis amplifier is 5 MHz. The x , y , and z amplifier characteristics of the Type 4501 make it particularly suited as a slave oscilloscope as these amplifier characteristics exceed all of the desirable criteria discussed earlier in this section. This system shown in Fig. 21-9 is self-explanatory; either X10 or X100 probes are used to couple the necessary signals from the master oscilloscope to the slave oscilloscope and, where necessary, Tektronix "input normalizers" are used to provide an additional 2X attenuation. The x and y signals are coupled differentially between the master oscilloscope and the slave oscilloscope and the differential z -axis amplifier is actually utilized as a summing amplifier to add together the master oscilloscope chopped blanking waveform and sweep retrace blanking waveform.

The Type 4501 Scan Converter Unit, as well as being basically a five-inch storage display unit, can internally scan information stored on the storage CRT and present this information on bright, large screen, TV monitors or receivers. The Type 4501 Scan Converter Unit is discussed in greater detail in Chapter 23.

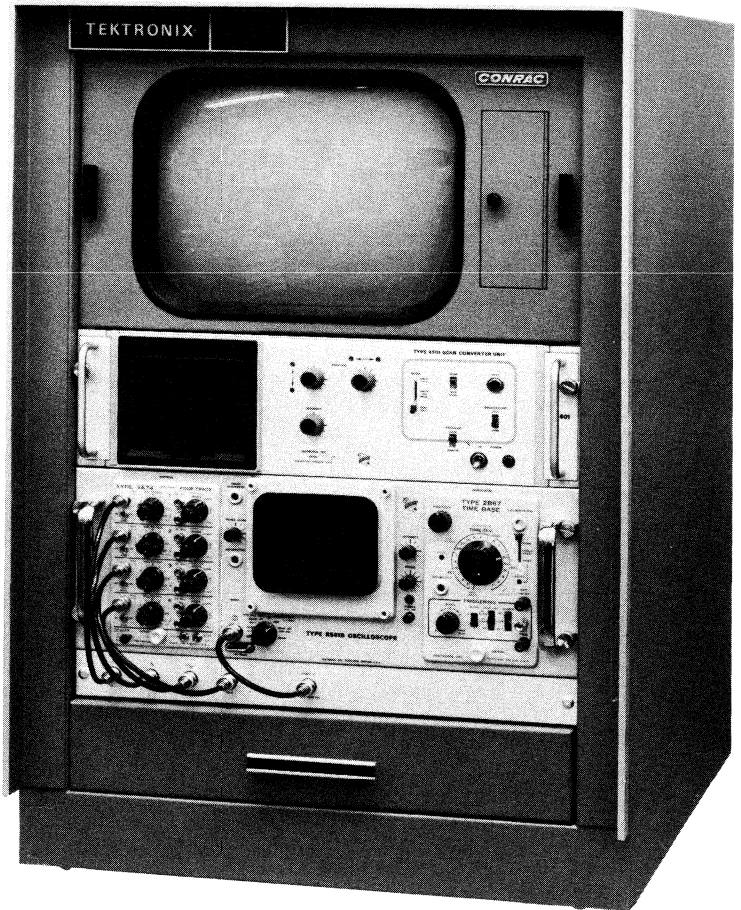


Fig. 21-10. A complete four-channel slave oscilloscope.

PULSE GENERATORS AND STIMULATORS

stimulation

As covered in various chapters in Section II of this book, the passage of electric current through various types of cells may cause these cells to depolarize and to generate an action potential. This passage of electric current is normally achieved biologically by a nerve impulse propagated from within the central nervous system; however, it may be artificially produced by a stimulator. The stimulator may be a self-contained instrument or it may be controlled by a separate pulse generator. For the purpose of this discussion of stimulus systems the pulse generator portion will be defined as that part that generates the pulse waveshape required for stimulation and the stimulator will be defined as that part that amplifies the signal to a level sufficient for artificial biological stimulation. Typically, a pulse generator may generate a 10 volt pulse; this pulse is then amplified within the stimulator to 100 volts or more for stimulation.

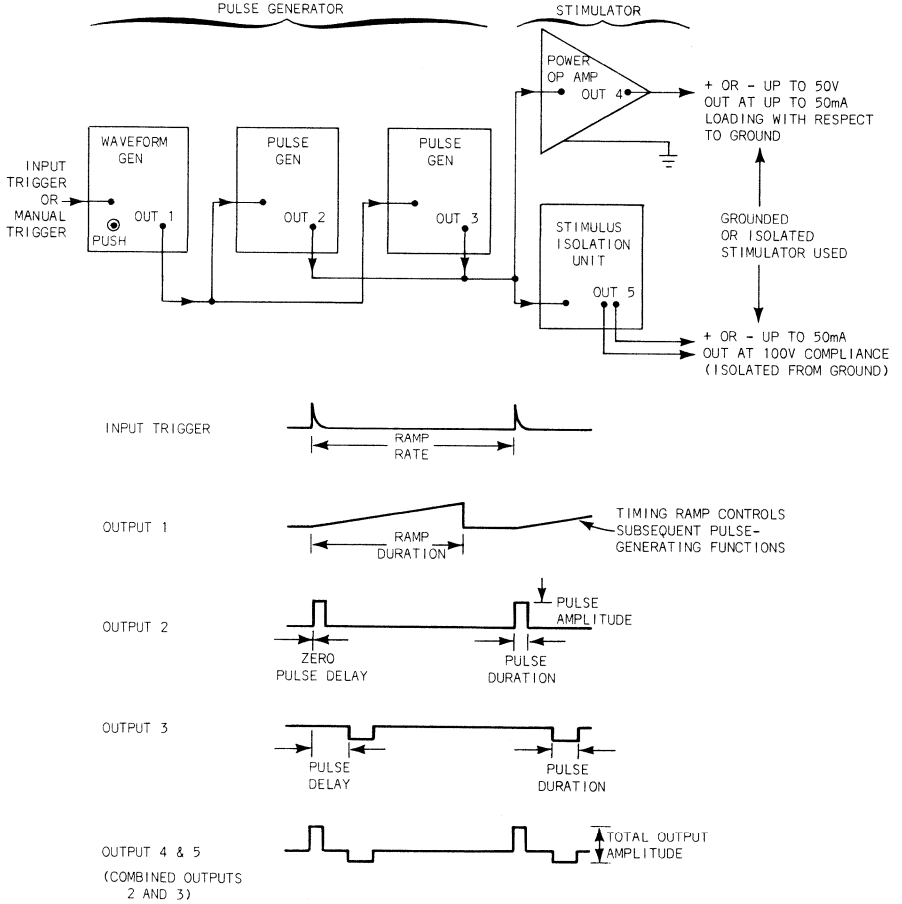


Fig. 22-1. A typical physiological stimulus system.

22.1 STIMULUS SYSTEMS

typical
system

A typical physiological stimulus system is shown in Fig. 22-1. The pulse generator section of this stimulus system uses a ramp generator to control two pulse generators. These pulse generators are triggered at discrete ramp levels, thus providing delay between the initiation of the ramp and the initiation of the pulse. With the system shown in Fig. 22-1, the ramp can either be generated repetitively, generated via an external trigger source or can be initiated manually from a push-button on the waveform generator. The output pulses from the pulse generators are combined to produce 2 positive pulses or a positive/negative pulse pair.

A stimulus system suited to most biophysical measurement requirements may have the following characteristics:

PULSE GENERATION

- Double pulse capability with each pulse independently controlled.
- Pulse widths from .01 ms to 300 ms.
- Pulse rise- and falltimes $< 1 \mu\text{s}$.
- Pulse repetition rate, controlled by ramp rate, from 1,000 Hz to 0.1 Hz with single pulse capability.
- Delay between pulses in a double-pulse format from ≈ 0 to 300 ms.
- Output pulses from constant current sources to allow mixing. Current amplitude and voltage compliance must be compatible with power operational amplifiers and stimulus isolation units. Usually 10 mA with compliance to 10 volts is adequate.

STIMULATION VIA POWER OPERATIONAL AMPLIFIER

- Input configuration to allow summation of either voltage source or current source outputs.
- Slew rate of $> 50 \text{ V}/\mu\text{s}$ with feedback elements adjusted for maximum output from a 1 volt or 10 mA input pulse.

- Up to ± 50 V/ ± 50 mA capability as either a voltage source or as a current source.

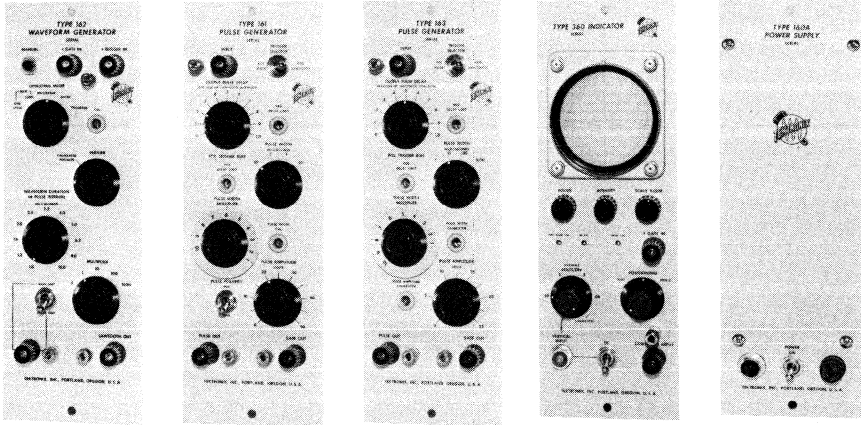
STIMULATION VIA STIMULUS ISOLATION UNIT

- Tristable output capability for stimulation with pulses having positive and negative components.
- Isolation between output and source of < 5 pF, $> 10^{10}\Omega$ and between output and ground of < 30 pF, $> 10^{10}\Omega$.
- Up to ± 100 V/ ± 50 mA capability as either a voltage source or as a current source.

brain stimulation Open cortical stimulation (stimulation of the exposed brain) requires considerably less stimulus energy than peripheral stimulation (stimulation of the arms and legs). Cortical stimulation requires from 0.1 to 10 V/0.1-10 mA while peripheral stimulation requires from 10 to 200 V/1-50 mA. Cortical stimulus energy should be maintained below 1 watt peak and 0.3 μ coulombs to avoid thermal and/or electrolytic injury to brain cells. Peripheral stimulation energy should be maintained below pain thresholds whenever possible.

22.2 TEKTRONIX 160 SERIES PULSE GENERATORS

Type 162 operation The Tektronix Type 162 Waveform Generator shown in Fig. 22-2 may be operated in one of three separate modes as shown in Fig. 22-3. When operating in the recurrent mode, the ramp will immediately reset itself to zero after completing a ramp cycle and begin a second ramp. This process will continue indefinitely. When the Type 162 is operated in the gated mode, the ramp generator will free-run for the duration of the input gate pulse, a ramp beginning when the input gate pulse goes positive and the last ramp being completed after the gate pulse again returns to zero. With the Type 162 operating in the triggered mode, the ramp generator may be triggered from external positive trigger pulses or by internal trigger pulses generated by manually depressing the trigger push-button on the front of the generator. One ramp is generated for each input trigger, thus, if only one input trigger is received, or if the front panel manual trigger push-button is depressed only once, a single ramp will be generated.



GENERATES A TIMING RAMP FROM 100μs TO 10s DURATION

GENERATES + OR - OUTPUT PULSE OF UP TO 50V WITH RISE- AND FALL-TIMES ≈ 1μs. DURATION VARIABLE 10μs TO 0.1s

GENERATES + OR - OUTPUT PULSE OF UP TO 25V WITH RISE- AND FALL-TIMES ≈ 0.2μs. DURATION VARIABLE 1μs TO 10ms

DISPLAY UNIT
Y = 0.05-50V/DIV
X = 11-15V/DIV
Z = 50V

POWER SUPPLY FOR 160 SERIES GENERATORS

Fig. 22-2. Tektronix "160" Series Pulse Generator system.

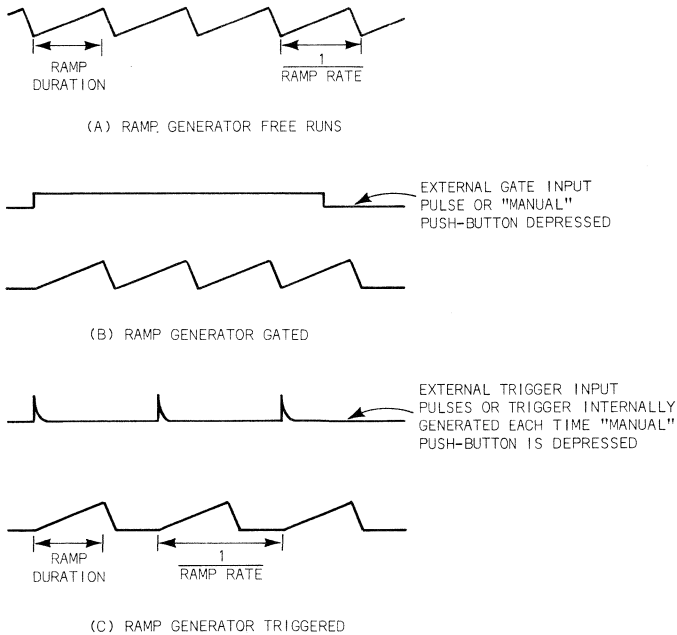


Fig. 22-3. Timing ramp control.

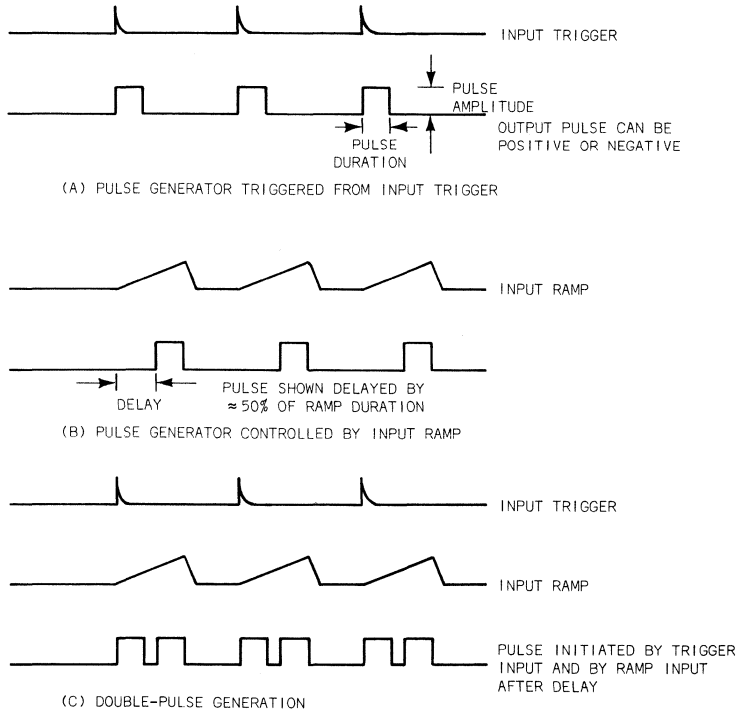


Fig. 22-4. Output pulse control.

Type 161 operation

The Type 161 Pulse Generator is shown in Fig. 22-2 and its various operating modes are shown in Fig. 22-4. The pulse generator may be triggered from an external positive trigger source, in which case the pulse output will begin when the trigger pulse goes positive. The pulse generation cycle may be initiated by an input ramp. Level-sensitive trigger circuitry within the pulse generator triggers at a precise level on this ramp and thereby provides a delay corresponding to the time taken for the ramp to reach the preset triggering level. The pulse generator may be triggered by both an input trigger and an input ramp, thus providing double pulse generation.

Types 163,
360 and
160A

The Type 163 Pulse Generator shown in Fig. 22-2 is similar to the Type 161 but provides a lower amplitude output pulse with faster rise- and falltimes. The Type 360 Indicator is basically a display device; its primary purpose being to display outputs from the 160 Series to simplify output pulse waveform adjustment. The Type 160A Power Supply provides regulated power to other 160 Series modules.

22.3 HIGH E OR I OUTPUT VIA A POWER OPERATIONAL AMPLIFIER

The output level from most pulse generators is insufficient for tissue stimulation, therefore, some form of pulse amplification is required between the pulse generator and the stimulating electrodes. A power operational amplifier, as discussed in Chapter 20, is ideally suited to this application.

constant
current
and
constant
voltage

Chapter 12, Section 12.1, discusses the characteristics of stimulators required for EMG use and points out that two different forms of stimulators are currently in use -- constant voltage stimulators and constant current stimulators. Although tissue can be stimulated with either a constant voltage source or a constant current source, it is difficult to correlate results obtained when using these different stimulating sources. Thus, while both systems are entirely adequate, the clinical electromyographer or physiologist will normally have a preference for either constant voltage or constant current stimulation, this preference probably being influenced by his earlier medical training.

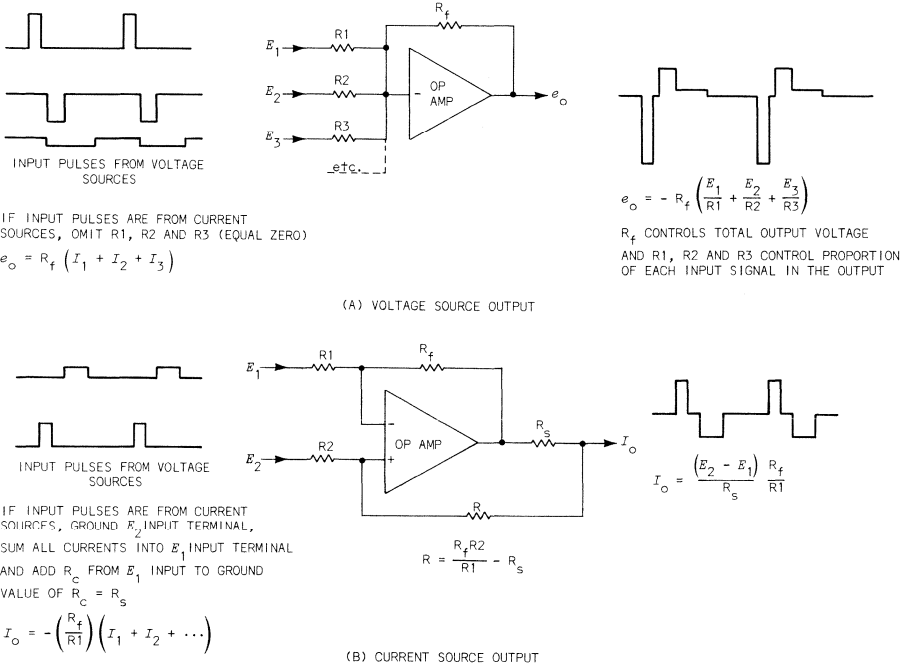


Fig. 22-5. Pulse combination and amplification with a power operational amplifier.

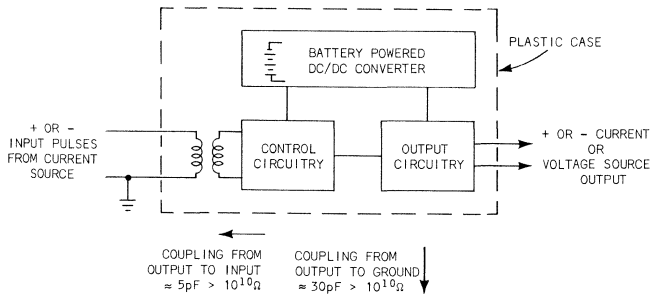


Fig 22-6. Equivalent circuit of a stimulus isolation unit.

constant
voltage

A power operational amplifier used in a voltage source configuration for constant voltage stimulation is shown in Fig. 22-5A. The input to the operational amplifier can be derived from either voltage sources or current sources and the output from the operational amplifier exhibits essentially constant voltage characteristics with an output impedance of considerably less than 10 ohms in most power operational amplifiers. Power operational amplifiers suitable for tissue stimulation should have an output capability of at least 50 V at 50 mA. It should be noted that the operational amplifier configuration shown in Fig. 22-5A inverts the output signal with respect to the input; in most instances, reversal of the stimulating electrodes effectively inverts the stimulating signal back to its original polarity.

constant
current

A power operational amplifier used in a current source configuration for constant current stimulation is shown in Fig. 22-5B. Since an operational amplifier inherently provides a voltage source output, a current sensing resistor (R_s) is used to detect the output current and to modify feedback around the operational amplifier to keep this current constant. Inputs from either voltage sources or current sources may be used. The output exhibiting constant current characteristics should have an output impedance typically in excess of 0.1 M Ω . An operational amplifier suitable for constant current stimulation should have an output capability of up to 50 mA with a voltage compliance to at least 50 V.

22.4 HIGH E OR I OUTPUT VIA A STIMULUS ISOLATION UNIT

character-
istics

A stimulus isolation unit provides amplification of an input pulse or pulses and isolates the output pulse from both ground and the input pulse source. The reasons for output pulse isolation are covered in detail in Chapter 12, Section 12.2. An equivalent circuit for a typical stimulus isolation unit is shown in Fig. 22-6. Since stimulus isolation may be used in applications requiring high amplitude stimulation into a relatively high impedance (such as in stimulation of the extremities using electrodes on the surface of the skin), the output from a stimulus isolation unit should have a greater voltage capability with a somewhat lower current capability than the output from a power operational amplifier.

A typical stimulus isolation unit operating in a constant voltage stimulation mode may produce a voltage output of up to 100 V with a current capability of 50 mA. Similarly, when operating in a constant current stimulation mode, it may provide a current output of up to 50 mA with compliance up to 100 V.

bistable
and
tristable

Stimulus isolation units are typically bistable or tristable devices. A bistable stimulus isolation unit offers either positive or negative outputs, having only two stable states -- output on and output off. A tristable stimulus isolation unit offers simultaneous positive and negative outputs, having three stable states -- output on positive, output off and output on negative. With a tristable stimulus isolation unit, the current or voltage provided by the "output-on-positive" pulse can be controlled independently from the current or voltage provided by the "output-on-negative" pulse.

22.5 CARDIAC PACEMAKERS

As discussed in Chapter 2 when dealing with the heart and the circulatory system, the steady rhythm of the heart is maintained by a biological "pacemaker" within the sinoatrial node of the heart. Failure of this pacemaker will cause the pumping action of the heart to be interfered with, causing a seizure and possible death.

Emergency resuscitation from a seizure can be accomplished by external electrical stimulation or

heart
stimulation

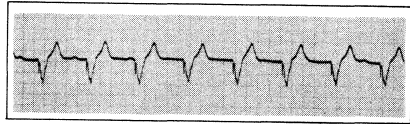
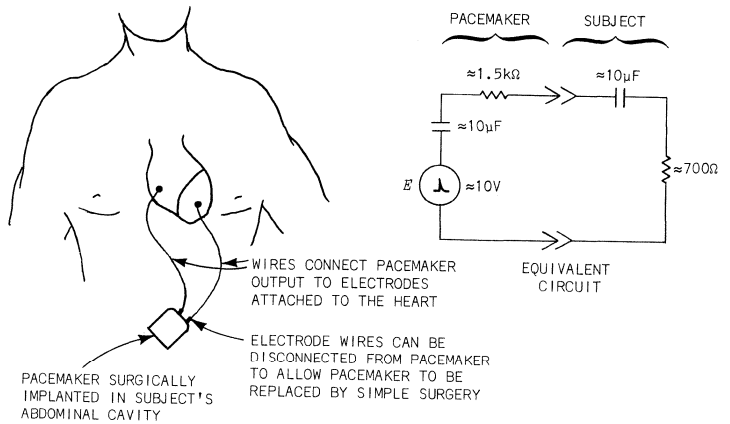
cardiac massage of the heart by trained medical personnel. Since in many subjects seizures are unpredictable, continuous stimulation is required to ensure continuous reliable operation of the heart. Long term stimulation from an external electrical source requires a considerable level of stimulating current to ensure that a sufficient level of current passes directly through the heart, thus causing considerable pain to the subject. This pain may be avoided by using external electrical stimulation via internal electrodes placed directly on the heart; however, tissue irritation and rejection present a problem that is difficult to overcome.

battery
life

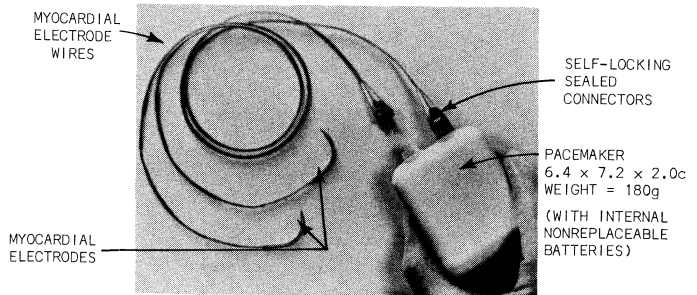
The modern cardiac pacemaker concept avoids tissue irritation and rejection by surgically implanting a pacemaker within the subject's abdomen and connecting it to the heart via internal electrodes. Modern internal pacemakers use mercury batteries which can operate the pacemaker continuously for a year or more. Since the pacemaker is placed in the subject's abdomen rather than in the heart, a relatively unsophisticated surgical procedure is necessary on a routine basis to replace the pacemaker with one having a fresh battery. The problem of determining just when an internal pacemaker's battery is nearly discharged, and thus, when the pacemaker must be replaced, has not yet been completely solved; however, many researchers have proposed various techniques for determining pacemaker end-of-life by analyzing the pacemaker output pulse as recorded on the surface of the body. At this stage there does not appear to be sufficient evidence in favor of any one method for it to be universally accepted.

pacemaker
output

The internal cardiac pacemaker consists of a transistorized blocking oscillator producing pulses of approximately 10 V in amplitude, a few milliseconds in width and at an approximate 60-per-minute rate. The equivalent circuit for a pacemaker and subject, together with a photograph of a commercial pacemaker, is shown in Fig. 22-7.



TYPICAL ECG FROM A SUBJECT WITH AN IMPLANTED CARDIAC PACEMAKER



A COMMERCIAL PACEMAKER PRODUCED BY ELECTRODYNE COMPANY INC., MASS., USA

Fig. 22-7. The implantable cardiac pacemaker.

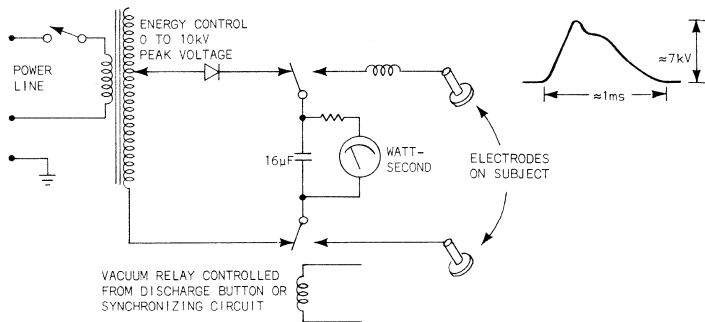
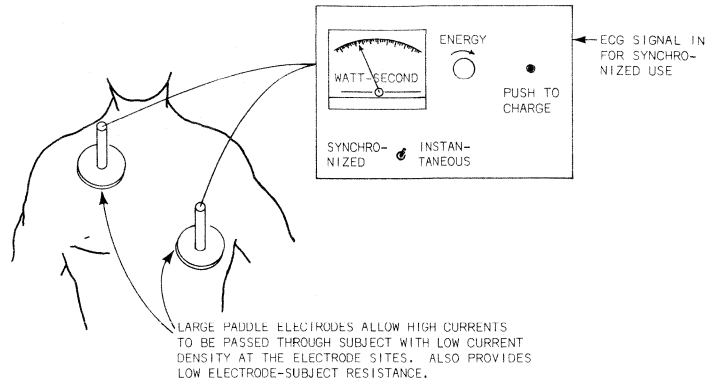


Fig. 22-8. The DC defibrillator.

22.6 CARDIAC DEFIBRILLATORS

ventricular fibrillation

Resuscitation from either a heart seizure or from ventricular fibrillation can be accomplished by external electrical stimulation. Ventricular fibrillation is produced within the human heart due to a variety of causes, including accidental electrocution. When the heart is in ventricular fibrillation, individual portions of the ventricular muscle contract independently instead of synchronously and effective output of blood ceases. External stimulation can be achieved with a cardiac defibrillator which essentially consists of a capacitor charged to several thousand volts which is then discharged through the subject via large-surface-area "paddle" electrodes as shown in Fig. 22-8. The energy produced within a cardiac defibrillator is normally measured in watt-seconds

watt-second shocks

or joules and is equal to $0.5 CE^2$, (C is the value of the storage capacitor and E is the voltage level to which it is charged). Defibrillation shocks from 200 to 400 watt-seconds are normally required to achieve satisfactory cardiac defibrillation. Not all of this energy will be available for dissipation at the subject as the efficiency of discharge will be considerably less than 100%; typically 20% to 70%.

instantaneous mode The cardiac defibrillator may be operated in one of two modes: instantaneous or synchronized. When operated in the instantaneous mode, the energy from the charged capacitor is discharged through the subject when the discharge button (usually located on one of the hand-held paddle electrodes) is depressed. It has been found that cardiac defibrillation is most efficient if the defibrillating pulse occurs during the falling part of the ECG R-wave and that defibrillation can be detrimental to some subjects if it occurs during the T-wave.

synchronized mode With a cardiac defibrillator operating in the synchronized mode, the discharge pulse is not immediately applied to the subject after the discharge button has been depressed but is delayed to occur during the falling part of the following R-wave. Use of a defibrillator in the synchronized mode necessitates an ECG signal being applied to the defibrillator for synchronizing purposes. Thus, a defibrillator can only be used in the synchronized mode if the ECG signal generated by the subject is of sufficient quality to allow the synchronizing circuit within the defibrillator to detect the R-wave. Many cardiac disorders produce abnormal ECG signals having detectable R-waves. These disorders may often be remedied by using a defibrillator in the synchronized mode, the defibrillator pulse *forcing* the heart to revert to a normal operating rhythm. This process is known as cardioversion

23

DISPLAY DEVICES AND INDICATORS

Both display devices and indicators are used to visually display the output from a biophysical instrumentation system. The display device utilizes a cathode-ray tube as the display medium, the indicator uses some other visual device such as a panel meter or a numerical readout device such as Nixie tubes. The following material discusses display devices from a biophysical measurements viewpoint only; therefore, we would recommend a companion volume entitled *Information Display Concepts* published by Tektronix, Inc. as a *general reference* source for display devices.

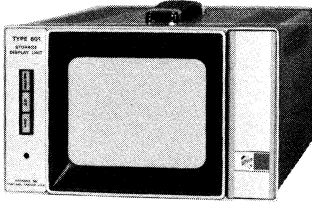
display
devices
versus
oscilloscopes

A display device may appear to be simply an oscilloscope, and indeed, an oscilloscope is often used as a display device. Display devices, however, generally differ from oscilloscopes as the CRT display within a display device is optimized for display characteristics (resolution, contrast, brightness and screen size) whereas, in contrast, the CRT display within an oscilloscope is optimized for measurement capability (vertical amplifier bandwidth and writing rate). A display device is intended to be used within a specific instrumentation system and thus has no external controls to change the characteristics of the display. An oscilloscope may, however, contain numerous external controls to change the characteristics of the display to suit the desired measurement requirement.

5in CRT UNITS

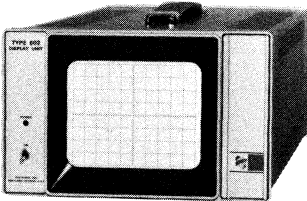
8cm x 10cm

TYPE 601 STORAGE DISPLAY UNIT



- x AND y - 1V FULL-SCREEN SENSITIVITY
- 100kHz BANDWIDTH
- 100k Ω 50pF INPUT RC
- z -AXIS - BISTABLE: ON > 1V OFF < 0.5V
- > 100kHz EFFECTIVE BANDWIDTH
- 100k Ω 50pF INPUT RC
- CRT - BISTABLE STORAGE
- 12.5 STORED LINE-PAIRS/cm
- RESOLUTION
- 5cm/ms WRITING SPEED

TYPE 602 DISPLAY UNIT

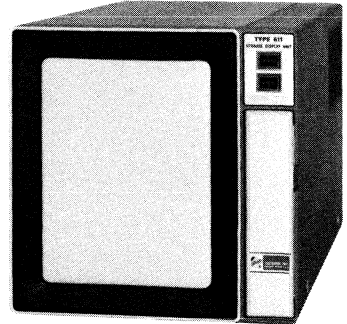


- x AND y - 1V FULL-SCREEN SENSITIVITY
- 1MHz BANDWIDTH
- 100k Ω 30pF INPUT RC
- z -AXIS - LINEAR: ON 1V RANGING TO OFF 0V
- 1MHz BANDWIDTH
- 100k Ω 70pF INPUT RC
- CRT - NONSTORAGE. P31 OR P7 PHOSPHOR
- < .014 INCH TRACE WIDTH

11in CRT UNIT

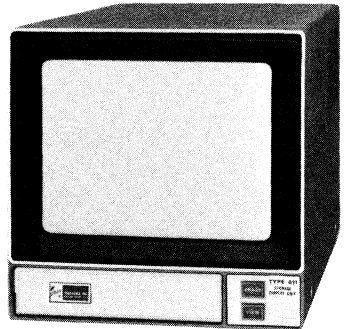
16.2 x 21cm

TYPE 611 STORAGE DISPLAY UNIT

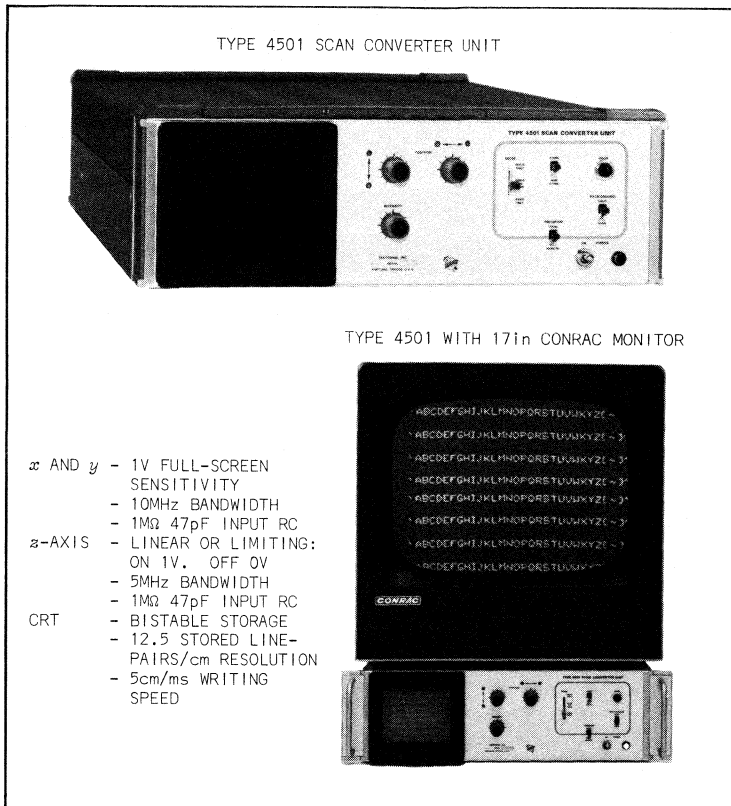


- x AND y - 1V FULL-SCREEN SENSITIVITY
- 100k Ω 60pF INPUT RC
- z -AXIS - BISTABLE: ON > 1V. OFF < 0.5V
- > 100kHz EFFECTIVE BANDWIDTH
- 100k Ω 50pF INPUT RC
- CRT - BISTABLE STORAGE
- 19 STORED LINE-PAIRS/cm
- RESOLUTION
- 25cm/ms WRITING SPEED

TYPE 611 MOD 162C - HORIZONTAL FORMAT



SCAN CONVERTER - 5in CRT TO 14in OR 17in TV MONITOR



(B)

Fig. 23-1. Tektronix display units.

23.1 TEKTRONIX DISPLAY UNITS

Tektronix currently produces four display units -- the Type 602 Display Unit, the Types 601 and 611 Storage Display Units and the Type 4501 Scan Converter Unit. These four products are shown in Fig. 23-1 together with a brief summary of their principle characteristics. All Tektronix display units require one volt of x and y input for full screen deflection horizontally and vertically and require one volt of z -axis input to turn the CRT beam on and off. As these x , y and z sensitivities are only adjustable over a relatively narrow range,

display
unit
character-
istics

it is necessary to externally amplify or attenuate signals to a level consistent with the sensitivity of the display unit before coupling such signals to the display unit. Tektronix display units may be used to provide an additional CRT display for a conventional oscilloscope, that is, in a slave oscilloscope configuration as discussed in Chapter 21.

Types 601
and 602

The Type 602 Display Unit utilizes a nonstorage, high resolution, CRT providing a display area of 8 cm vertically by 10 cm horizontally. The Type 601 Storage Display Unit has similar characteristics to Type 602 Display Unit and incorporates a storage CRT to allow displayed information to be stored as discussed in Chapter 21. Erasure of stored information can either be accomplished manually with a push-button on the front of the Type 601 or may be accomplished by an electrical signal.

Type 611

The Type 611 Storage Display Unit provides four times the display area of the Type 601 by utilizing an 11-inch storage CRT. The display area of the Type 611 is 16.2 cm horizontally by 21 cm vertically; the instrument is also available in a horizontal format providing 16.2 cm vertically and 21 cm horizontally. A stored display on the Type 611 may be either erased by a push-button or remotely by an electrical signal.

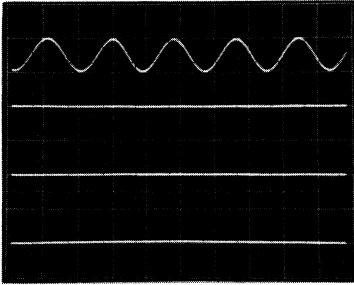
Type 4501

The Type 4501 Scan Converter Unit (see Fig. 23-1B) is a unique instrument providing a large screen CRT display via a television system. The Type 4501 Scan Converter may basically be considered as a 5-inch storage display unit (similar to the Type 601) coupled with a television system to display information stored on the scan converter's CRT on a large screen television monitor or a commercial television receiver. The bright displays achieved via scan conversion are ideal for individual or group viewing under high ambient light conditions.

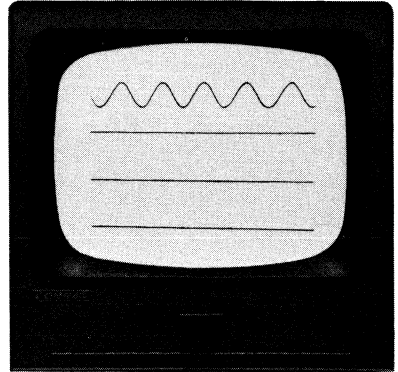
Readout from the storage CRT is accomplished by scanning the CRT storage target with a TV raster. All necessary scanning, sync, and video circuits are contained in the scan converter, providing a composite TV signal of the information stored on the CRT.

The TV signal is available as either EIA 525-line, 60-field format or in CCIR 625-line, 50-field format at 1 volt P-P. This signal can be used in conjunction with a studio-quality TV monitor for large screen display. Modulated RF at 55.25 MHz through 67.25 MHz is also produced to allow the use of commercial TV receivers for a display.

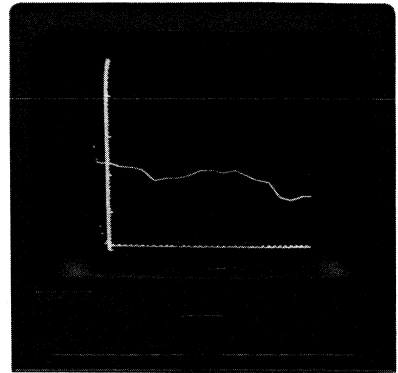
STORED DISPLAY
ON SCAN CONVERTER CRT



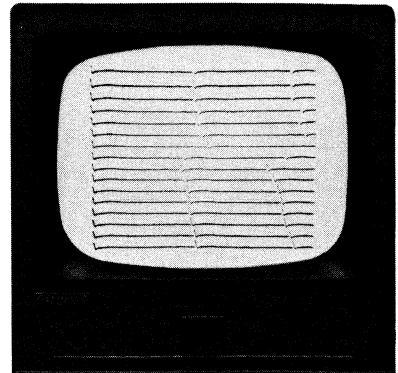
DISPLAY MAY BE REPRODUCED TO
ANY SIZE VIA TV MONITOR
PROJECTION SYSTEM OR TV



MULTI-CHANNEL OSCILLOSCOPE DISPLAYS



COMPUTER GENERATED DISPLAYS



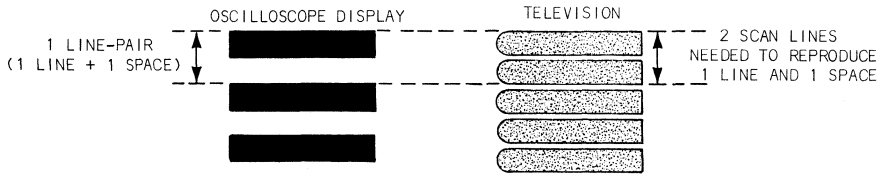
SUBJECT ECG OVER ONE MINUTE
(REFER TO FIG. 23-4)

Fig. 23-2. Type 4501 Scan Converter capabilities.

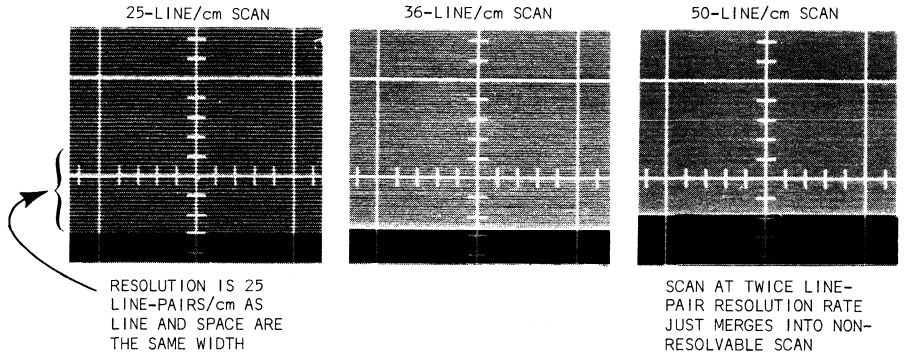
variety of
information

As shown in Fig. 23-2, the Type 4501 Scan Converter or any other display unit can be used to present many different types of information. The Type 4501 Scan Converter's application as a slave oscilloscope is discussed in Chapter 21. A typical four-channel slave oscilloscope display on a television monitor is shown in Fig. 23-2 together with a computer generated display of a subject's variation in temperature over a 36-hour period and a stepped-trace display of a subject's ECG measured over a one-minute period using the system discussed later in this chapter.

THE RESOLUTION OF AN OSCILLOSCOPE DISPLAY IS MEASURED IN LINE-PAIRS AS SHOWN



SCAN ON A TYPE 602 DISPLAY UNIT CRT SHOWING LINE-PAIR RESOLUTION AND EFFECT OF CLOSER SCANNING



TOP LEFT-HAND QUADRANT OF HIGH RESOLUTION DISPLAYS ON TEKTRONIX DISPLAY UNITS SHOWING RESOLUTION UNIFORMITY

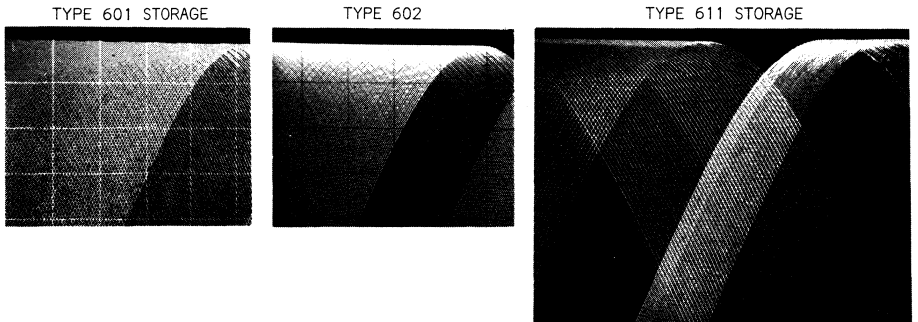


Fig. 23-3. Resolution of Tektronix display units.

23.2 RESOLUTION

line-pairs
versus
scan lines

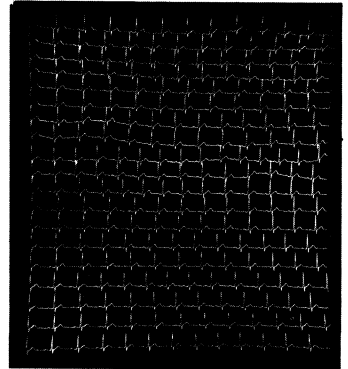
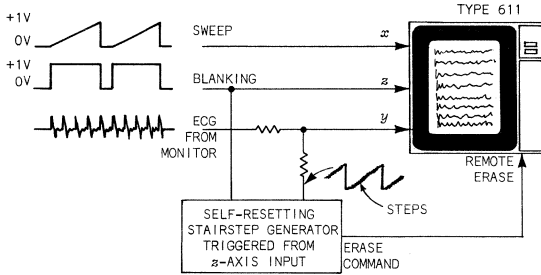
The principal difference between the CRT display on an oscilloscope and a display unit is the increased brightness and resolution obtainable with a display unit as shown in Fig. 23-3. The resolution of a display unit or oscilloscope CRT display is specified in line-pairs, one line-pair being equal to one bright written line and one space of equal width. This specification should not be confused with scanning resolution specified in *lines* as associated with television systems; in such systems two scan lines are required to produce one line and one space.

determining
resolution

A scan of 25 lines/cm on a Type 602 Display Unit is shown in Fig. 23-3. From an observation of this display it appears that *the width of the dark space between these lines is equal to the width of the lines*, thus it can be said that this display has a resolution of 25 line-pairs/cm. If the scanning density is now increased to 36 lines/cm as shown on Fig. 23-3, the individual scanning lines can still be clearly discerned; however, it is apparent that the width of the scanning lines is somewhat greater than the width of the dark space between the scanning lines. At a scanning rate of 50 lines/cm the dark space between the scanning lines begins to disappear and the individual scanning lines begin to merge into a nonresolvable scan. The resolution of this particular CRT could be quoted as 25 line-pairs/cm or, when dealing with a television type display, as having a resolution capability of 50 lines/cm.

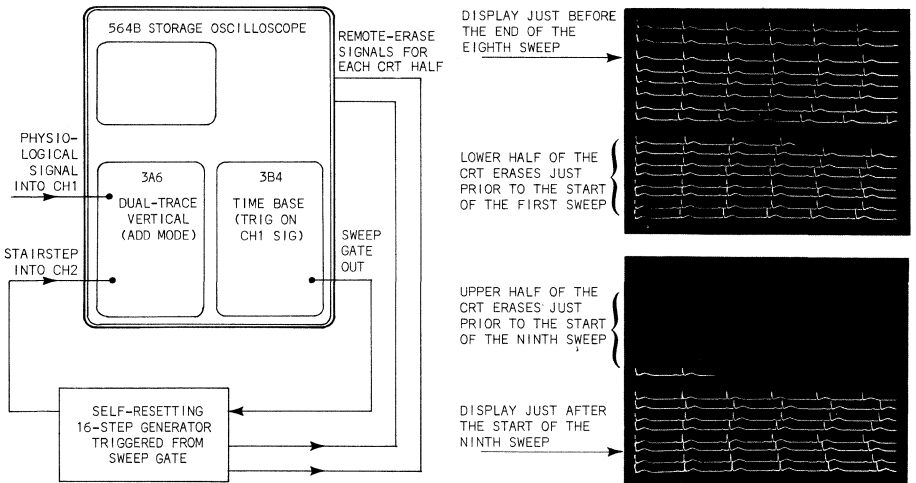
spot size

The spot size of the before referenced display is calculated to be .008 inches as the display is capable of 25 line-pairs/cm, that is, 50 spots/cm. It should be noted that the spot size specified for the Tektronix Type 602 Display Unit is .014 inches, almost twice the spot size calculated above. Spot size will vary from one CRT to another; it will also vary slightly over the total display area of the CRT and varies considerably with the intensity of the display. Although the typical spot size for a Type 602 Display Unit may be .008 inches, the specification of .014 inches represents the maximum spot size that would be encountered at any point on any Type 602 CRT when displayed at a relatively high intensity level.



ECG DISPLAY OVER A 3-MINUTE PERIOD

(A) USING THE TYPE 611 IN A SYSTEM



REFER TO CHAPTER 27 FOR DETAILS OF THE STEP GENERATORS

(B) SELF-CONTAINED SYSTEM USING A TYPE 564B SPLIT-SCREEN STORAGE OSCILLOSCOPE

Fig. 23-4. Expanded x-axis storage-CRT displays.

The uniformity of resolution between the center of a CRT display and the corners of the display is particularly good with Tektronix display units as shown in the lower series of photographs in Fig. 23-3. An observation of these photographs will show that little loss of resolution can be detected when comparing the resolution at the center of the CRT display to the resolution at the top left-hand edge of the CRT display.

23.3 LARGE SCREEN EXPANDED-SWEEP DISPLAYS

The normal 5-inch CRT of most oscilloscopes limits the amount of information that may be presented on the CRT without confusion. The large screen capabilities of the Type 611 Storage Display Unit and the Type 4501 Scan Converter Unit allow a greater quantity of information to be displayed. This greater quantity of information may be derived from many different channels of information or may be derived from a single channel of information over a prolonged period of time using vertical sweep stepping as shown in Fig. 23-4.

sweep
stepping

The displays shown in Fig. 23-4 require the use of a step generator capable of being updated with each successive sweep. The step generator may also contain an adjustment to allow the number of sweeps to be varied to suit the display requirements. Details of two self-resetting stair-step generators are given in Chapter 29.

expanded
 x axis

The expanded x -axis display shown in Fig. 23-4A uses the Type 611 to store a large volume of analog information on its 11-inch CRT. Although the display shown represents a subject's ECG history over a three minute period, the resolution of the Type 611 is such as to allow similar displays to be presented over far greater periods by using slower sweep speeds, a lower level of ECG signal and less spacing between sweeps. An erase command from the step generator will erase the CRT when the step generator resets to zero to once again begin its stepping cycle or, alternatively, the information may be held on the CRT for a prolonged period or until the step generator is manually reset to erase the information and begin the cycle once again. While this system has the advantage of a large screen display, all of the information stored on the screen must be erased

signal
density

before beginning a new series of sweeps, thus, the amount of "history" presented on the CRT varies from zero just after erasure to three minutes, as with the display shown in Fig. 23-4A when the step generator has almost completed its cycle and the screen is full of information. This disadvantage is overcome in the system shown in Fig. 23-4B.

The expanded x -axis storage display shown in Fig. 23-4B uses a Type 564B Storage Oscilloscope with a Type 3A6 vertical plug-in unit and a Type 3B4 horizontal plug-in unit in conjunction with a self-resetting 16-step generator. Although this system only uses a 5-inch CRT display, it has the advantage of being a self-contained system and, as described below, uses the split-screen storage feature of the Type 564B to alternately erase segments of the CRT.

The expanded x -axis display system shown in Fig. 23-4B provides 16 sweeps, the first sweep beginning at the bottom left-hand corner of the CRT and the last sweep finishing at the top right-hand corner of the CRT. Just prior to the completion of the last sweep the CRT will present the previous 16 sweeps of stored information. At the end of the last sweep the step generator will reset to zero and the lower half of the CRT will be erased, thus leaving only the previous 8 sweeps presented on the upper half of the CRT. The system will then commence to cycle through its first 8 sweeps, storing them on the lower half of the CRT. At the end of the eighth sweep the step generator will cause the upper half of the CRT to be erased, leaving only the information generated by the previous 8 sweeps on the lower half of the CRT. The step generator will then commence to cycle through the remaining 8 sweeps, storing information on the upper half of the CRT. The cycle is then repeated continuously. This system has the advantage that at least 8 sweeps are always presented on the storage CRT at any one time, thus the amount of "history" presented at any one time varies from 8 sweeps to 16 sweeps rather than, as with a system using a nonsplit-screen storage oscilloscope, from zero to 16 sweeps.

programmed
erase

23.4 INDICATORS

moving-
coil meter

As stated previously, indicators may consist of a moving-coil panel meter or some form of digital display device. The moving-coil panel meter, while being considerably less expensive than a digital indicating device, is somewhat difficult to read and thus should only be used for the indication of noncritical parameters or where digital display devices would be uneconomical. Moving-coil meters may be directly coupled to other components in a biophysical monitoring system or may be coupled to these components via amplification circuits.

digital
meters and
indicators

Digital meters and indicators accept voltage levels from other components in a biophysical monitoring system and convert them to a format suitable for display in numeric form. They offer the advantage of showing the measured quantity in a form that can be read at a glance or at a considerable distance. The digital indicator is also free of any reading error that may occur when reading a moving-coil pointer against a fixed scale. Digital display indicators are generally custom built for specific systems. Electronics for Medicine, New York, produces a digital display unit for use in an operating room providing 6 channels of numeric display to indicate the subject's temperature, pulse rate, venous pressure, arterial diastolic pressure, arterial systolic pressure and mean arterial pressure. A numeric indicator device may consist of a CRT display unit and a character generator for characters on the CRT. The character generator may be a separate instrument or may utilize a digital computer to generate characters with the aid of digital to analog converters.

Digital displays are not the whole answer to data presentation. Particularly if the user is harassed or tired, they are easily misinterpreted. Further, they give no indication of rate-of-change of the variable shown, and the human brain seems to be able to accept rate-of-change information separately from absolute levels. It is far easier for the human brain to accept a pattern than a digital value.

The best way of looking at the matter is to ask how many bits of information, or information levels, are actually required. Is it just "Too much" or "Too little," or is it "Much too much," "Too much," "O.K.," "Too little" or "Much too little?" A level display which may include absolute values for recording or closer examination if time permits is far more valuable than just numbers.

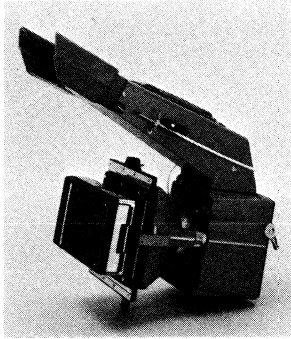
OSCILLOSCOPE CAMERAS

Biophysical information displayed on an oscilloscope or display unit must be photographically recorded if a permanent record of this information is to be maintained. While a storage oscilloscope does maintain a record for an hour or more, it too must be photographed if a more permanent record is required.

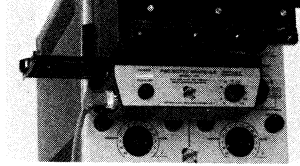
Information displayed on a CRT may be photographed using either conventional photographic techniques or moving-film photographic techniques. Conventional photography of information displayed on a CRT involves photographing the CRT display with an oscilloscope camera. While this technique is entirely satisfactory for most applications, it does not provide a continuous record of data versus time, but only a record of data over a discrete period of time (normally the duration of one CRT horizontal sweep). Continuous data may be recorded by photographing a CRT display with a movie camera; however, the movie film can only be displayed by projection and does not provide a convenient record that can be studied with ease. A modification of movie photography, known as continuous-motion photography, provides a permanent record of data versus time on photographic film by eliminating the horizontal sweep from an oscilloscope and by providing the horizontal "sweep" by moving the film past the face of the CRT as discussed in Section 24.3.

continuous-
motion
photography

TYPE C-12 CAMERA



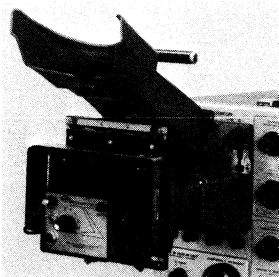
PROJECTED GRATICULE ACCESSORY FOR TYPE C-12 CAMERA



PROJECTED GRATICULE ALLOWS SPECIAL GRATICULES OR HAND WRITTEN DATA TO BE SUPERIMPOSED ONTO THE FILM TOGETHER WITH DATA FROM THE CRT.

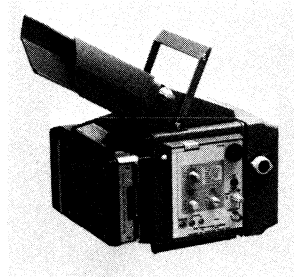
BEAM-SPLITTING MIRROR MINIMIZES VIEWING PARALLAX WITH CRTS HAVING EXTERNAL GRATICULES. ACCEPTS A WIDE RANGE OF LENSES AND FILM BACKS.

TYPE C-27 CAMERA



DIRECT VIEWING WITHOUT THE USE OF A MIRROR, THUS TRANSMITTING MAXIMUM LIGHT TO THE FILM. ACCEPTS A WIDE RANGE OF LENSES AND FILM BACKS.

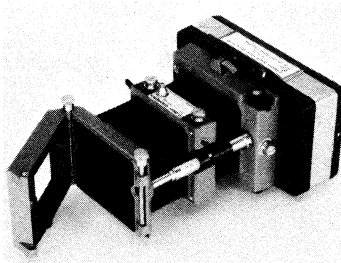
TYPE C-50 CAMERA



F1.9 LENS 1:0.7 MAGNIFICATION EASE OF OPERATION AFFORDED BY BUILT IN TRACE-BRIGHTNESS PHOTOMETER, RANGE-FINDER FOCUSING AND ACCURATE EXPOSURE CONTROL.

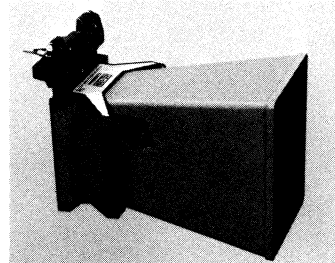
(A) FOR SIMULTANEOUS PHOTOGRAPHY AND VIEWING - FOR USE WITH MOST TEKTRONIX OSCILLOSCOPES WITH 5in CRTS

TYPE C-30A CAMERA



FOR USE WITH TEKTRONIX OSCILLOSCOPES HAVING 3, 4, OR 5in CRTS. F1.9 LENS MAGNIFICATION VARIABLE IN INDEXED STEPS FROM 1:0.7 TO 1:1.5. ACCEPTS A WIDE RANGE OF OPTIONAL FILM BACKS.

TYPE C-10 CAMERA



FOR USE WITH TEKTRONIX OSCILLOSCOPES OR MONITORS HAVING 11in CRTS F8 LENS - FIXED FOCUS - 1:0.5 MAGNIFICATION

(B) FOR PHOTOGRAPHY OR VIEWING WITH CAMERA REMOVED OR "SWUNG AWAY"

Fig. 24-1. Tektronix oscilloscope cameras suited to biophysical measurements.

24.1 CONVENTIONAL CRT PHOTOGRAPHY

Conventional oscilloscope photography involves taking a still photograph of the CRT. Specialized cameras are, however, required to eliminate extraneous light from the photographic record and to allow the CRT display to be photographed without distortion by a lens with a relatively short focal length. Since it is usually undesirable to wait for a film to be processed before being able to analyze data recorded on it, CRT cameras normally utilize Polaroid* film, providing a permanent record within 10 seconds. Polaroid film is produced by the Polaroid Corporation, Massachusetts. It is often desirable for an oscilloscope camera to incorporate a viewing tunnel or some other form of viewing mechanism to allow the CRT display to be viewed during photography.

Type C-12
with beam-
splitting
mirror

The Tektronix Type C-12 Camera shown in Fig. 24-1 uses a beam-splitting mirror between the lens and the CRT. This mirror permits part of the light from the CRT to pass through the mirror to the lens and part of the light from the CRT to be reflected by the mirror into the viewing tunnel. This permits viewing from an effective viewing position directly in front of the CRT, thus minimizing parallax error between an external graticule and the information displayed on the CRT. The beam-splitting mirror within the Type C-12 Camera also allows light projected from beneath the camera to be reflected into the lens and to pass through the mirror into the viewing system. A projected graticule accessory for the Type C-12 Camera provides a light source beneath the camera and allows various special graticules and masks to be inserted between this light source and the camera. The projected graticule accessory eliminates any parallax from an external graticule CRT by providing a supplementary graticule in the same plane as the CRT phosphor. Many specialized graticule formats are available from Tektronix for use with the projected graticule; most of these special graticules include a clear area to allow the user to write information on the graticule which will then appear on the final photographic record.

projected
graticule

*Trademark Polaroid Corporation

- Type C-27 Although the Tektronix Type C-27 Camera (Fig. 24-1) is similar to the Type C-12 Camera, it incorporates direct angular viewing of the CRT rather than a beam-splitting mirror. This permits the maximum transmission of light from the CRT to the lens. As it does not include a beam-splitting mirror, it does not allow the use of the projected graticule accessory. The Type C-27 Camera shown includes a speed computer and an electric shutter; either manual shutters or electric shutters are available on many Tektronix oscilloscope cameras.
- Type C-50 The Type C-50 Camera (Fig. 24-1) is particularly easy to operate due to the inclusion of a built-in trace brightness photometer, a range finder focusing mechanism and an accurate exposure control system. The Type C-50 provides angular viewing of the CRT and does not include a mirror between the CRT and the lens.
- Type C-30A The Type C-30A Camera (Fig. 24-1) does not include provision for simultaneous photography and viewing; it must be either removed from the oscilloscope for viewing or it may be swung away to one side via its hinged mounting bezel. The Type C-30A Camera is particularly versatile, being suited for use with most Tektronix oscilloscopes and including bellows between the lens, CRT plane and film plane to allow variable magnification from 1:0.7 to 1:1.5.
- Type C-10 The Type C-10 Camera (Fig. 24-1) is intended for use with Tektronix oscilloscopes or monitors having 11-inch CRT's, such as the Type 611 Display Unit or the Type T4002 Graphic Computer Terminal.
- film backs All of the Tektronix oscilloscope cameras shown in Fig. 24-1, with the exception of the Type C-10, are shown with film backs for use with Polaroid Type 107 Pack Film, providing "instant" photographic records on 3 1/4-inch X 4 1/4-inch Polaroid film. All of these cameras can be used with Graflok* film holders and accessories to allow the camera to be used with cut-film holders, film-pack adapters, roll-film (120) holders, etc. Graflok backs and accessories are available from local camera shops. The Type C-10 Trace-Recording Camera incorporates a Graflok back to allow its use with Polaroid Type 57 4 by 5-inch cut film.

*Trademark Graflex, Inc.

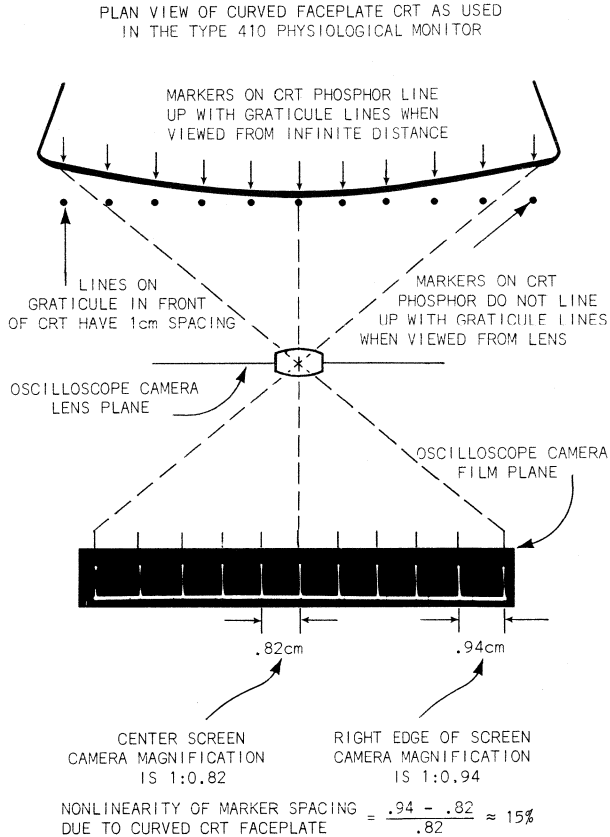
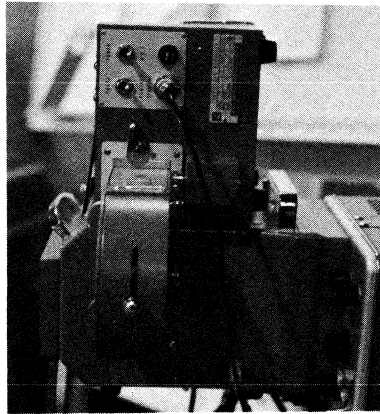


Fig. 24-2. Photography of a curved faceplate CRT.

24.2 PHOTOGRAPHY OF CURVED FACEPLATE CRT'S

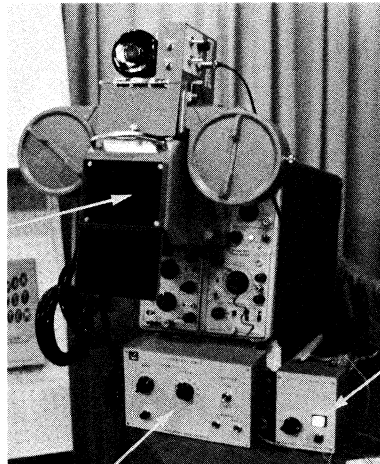
curved
faceplates

All Tektronix oscilloscope cameras are intended for use with CRT's having flat faceplates, the lenses are thus compensated for good linearity when used with flat-faceplate CRT's. Some CRT's, such as the CRT incorporated within the Tektronix Type 410 Physiological Monitor, have curved faceplates. A curved-faceplate CRT such as the Type 410 CRT will produce an approximate 15% nonlinearity in the photographic record when photographed with a conventional oscilloscope camera as shown in Fig. 24-2.



THE NIHON KOHDEN CAMERA MOUNTS DIRECTLY ON MOST TEKTRONIX OSCILLOSCOPES, THUS ELIMINATING THE NEED FOR CUSTOM ADAPTERS

OPENS FOR SIMULTANEOUS PHOTOGRAPHY AND CRT VIEWING BY OPERATOR



ILLUMINATES A DATA CARD FOR RECORDING PERMANENT DATA ON THE FILM

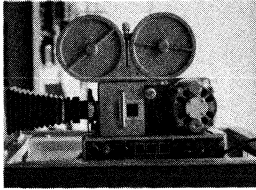
CONTROL UNIT SELECTS CONTINUOUS MOTION OR SINGLE FRAME OPERATION AND FILM SPEED FROM 100cm/s TO .05mm/s

Fig. 24-3. The Lehigh Valley Electronics, Inc., Nihon Kohden PC-2A continuous-motion camera.

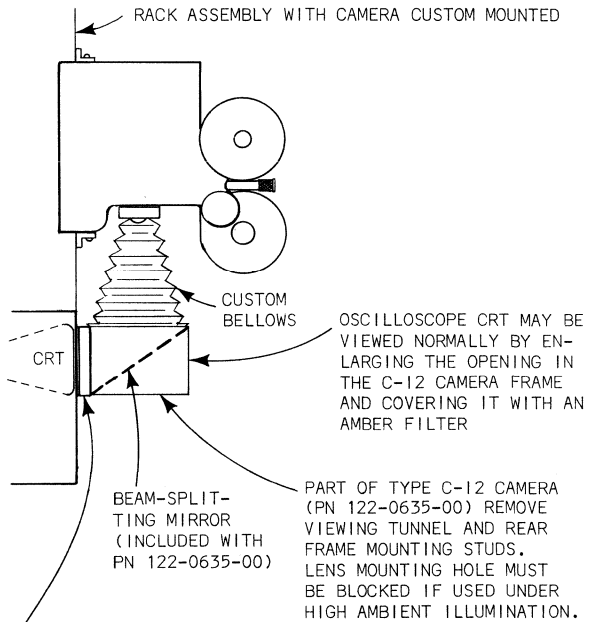
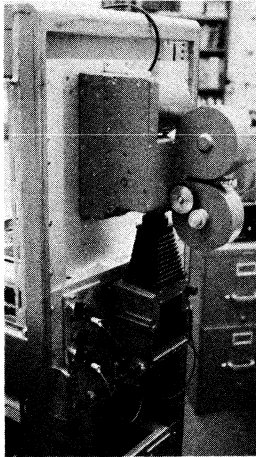
24.3 CONTINUOUS MOTION CAMERAS

A continuous-motion camera provides continuous motion of film or sensitized paper (usually 35 mm) past an oscilloscope CRT via a lens system. Since horizontal information is provided by the moving film, the horizontal sweep of the oscilloscope display is turned off leaving only vertical information deflecting the CRT beam. This recording system can be likened to a conventional chart recorder with vertical movement of the CRT beam acting in a similar manner to vertical movement of the stylus on the chart recorder. Two commercial continuous motion cameras are shown in Figs. 24-3 and 24-4. The Nihon Kohden Model PC-2A Continuous Motion Camera shown in Fig. 24-3 is marketed in the USA by Lehigh Valley Electronics, Inc. The camera is produced in Japan and is available with a full range of adapters to allow the camera to be conveniently mounted on most Tektronix oscilloscopes. A broad range of film transport speeds is provided from 1000 millimeters per second to 0.05 millimeters per second.

Nihon
camera



HORIZONTAL MOUNTING IN FRONT OF CONVENTIONAL OSCILLOSCOPE



TEKTRONIX OSCILLOSCOPE/CAMERA ADAPTER
 PN 016-0226-01 FOR MOST 5-INCH
 ROUND CRT'S
 PN 016-0217-01 FOR MOST 5-INCH
 RECTANGULAR CRT'S
 REFER TO CURRENT TEKTRONIX
 CATALOG FOR PN VERIFICATION

Fig. 24-4. The Grass Instrument Company, Kymograph continuous-motion camera.

Kymograph
camera

The Kymograph* Continuous Motion Camera shown in Fig. 24-4 is manufactured by the Grass Instrument Company in the United States. Film transport speeds from 1000 millimeters per second to 0.25 millimeters per second are provided. The Grass Instrument Company does not produce adapters to allow this continuous motion camera to be easily adaptable to most commercial oscilloscopes, thus some form of custom mounting, either in a horizontal or vertical format, must be developed as shown in Fig. 24-4. This custom mounting may be achieved using part of the Tektronix Type C-12 Camera to provide a beam-splitting mirror to allow the CRT to be viewed during continuous motion photography. An amber filter must be incorporated into the viewing path to block out all but amber extraneous light from entering the camera lens via the viewing path. Most recording films are blue sensitive, thus the amber light will not expose the film. If, however, panchromatic film is used, then the amber extraneous light can be rejected with a blue filter in front of the camera lens. The Nihon Kohden camera incorporates a beam-splitting mirror and filters for simultaneous viewing and photography. Both the Nihon Kohden and Grass continuous motion cameras can also be used for conventional, single-picture oscilloscope photography with the oscilloscope time base operating.

If either camera (incorporating a 110 volt motor) is to be used on a higher voltage supply by using a step-down transformer, the transformer should be of adequate rating (500 volt-amp or more) to deliver the *starting surge* current of the motor.

*Trademark Grass Instrument Company

25

GRAPHIC RECORDERS

A continuous visual record of physiological data versus time may either be obtained with a graphic recorder or with an oscilloscope and a continuous motion camera as covered in the previous chapter. Continuous motion CRT photography, while offering almost unlimited bandwidth and extremely fast recording rates, does not provide an immediate record of the recorded data. A graphic recorder, while inherently possessing limited high-frequency response characteristics, does provide an immediate record of the recorded data for observation. Graphic recorders are often referred to as oscillographs, oscillographic recorders, strip chart recorders or chart recorders. Graphic recorders utilize some form of stylus to traverse a strip of chart paper while the chart paper is in motion, leaving a visible tracing which is a record of the time variations of the input voltage or current.

Most graphic recorders consist of the following five basic components:

basic
components

- An electromechanical device to convert an electrical input signal to mechanical movement,
- A stylus arm to transmit the mechanical movement from the electromechanical device to the stylus,
- A stylus to leave a written record on chart paper as the stylus moves across the chart paper,
- A chart paper assembly consisting of a chart paper supply roll, a chart paper writing table and a chart paper takeup roll and
- A paper drive mechanism to move the chart paper across the writing table from the supply roll to the takeup roll at a constant speed.

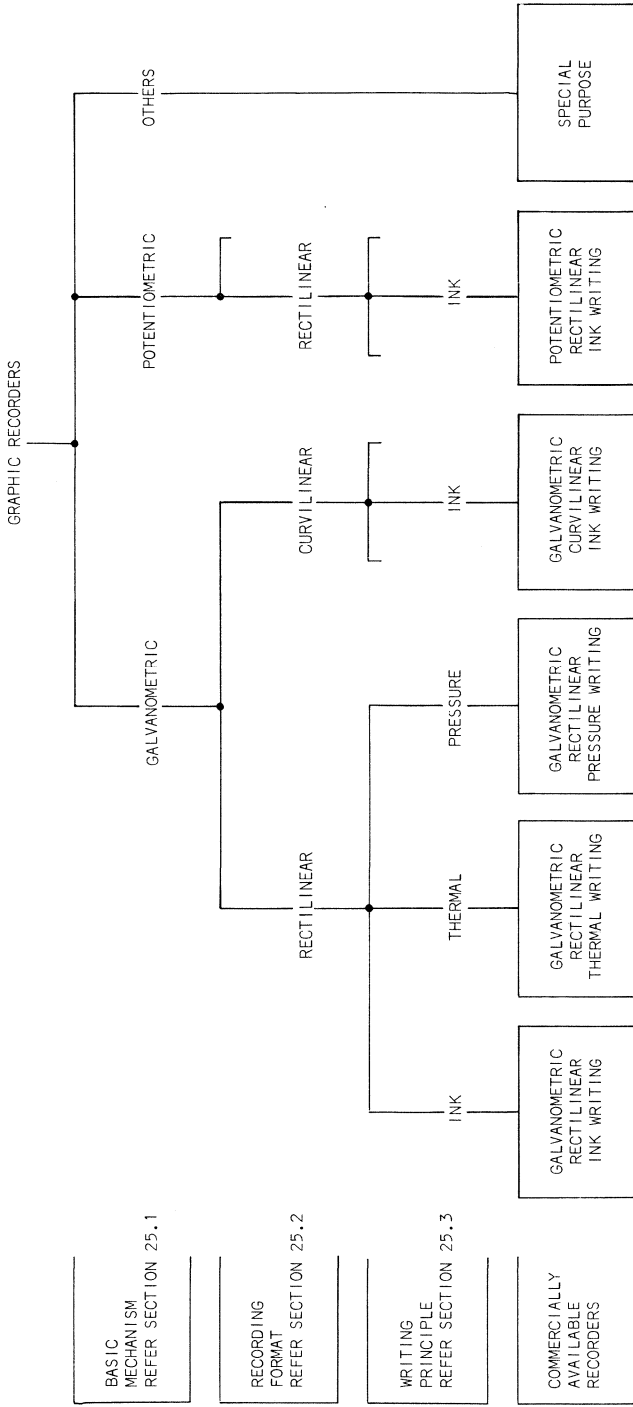


Fig. 25-1. Graphic recorder classifications.

character-
istics

Graphic recorders may be broadly classified as to the following three characteristics:

-- Basic Mechanism

The basic mechanism used to convert an input current or voltage into mechanical movement of the recording stylus may use either a galvanometric or a potentiometric principle as discussed in Section 25.1.

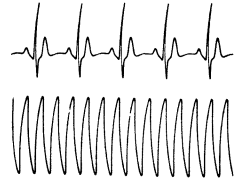
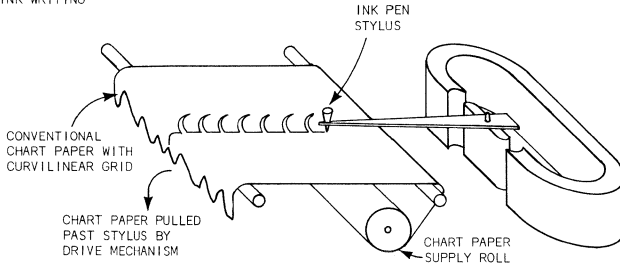
-- Recording Format

The geometrical relationship between stylus movement and the chart paper will either result in a curved line or a straight line being transcribed on the chart paper when the stylus is abruptly moved to a new position (referred to as curvilinear or rectilinear recording respectively as discussed in Section 25.2).

-- Writing Principle

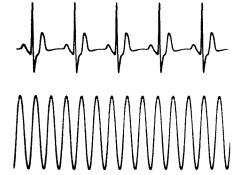
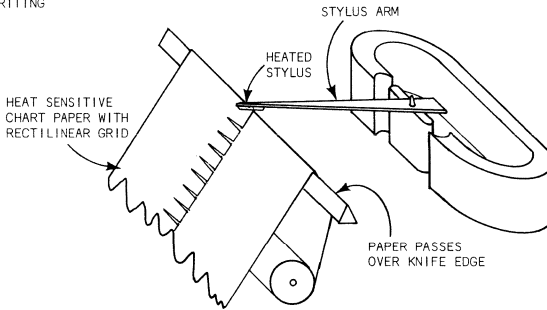
A written record is transcribed on the chart paper with either an ink-pen stylus, a heated stylus or a rounded-point stylus (referred to as ink writing, thermal writing and pressure writing respectively as discussed in Section 25.3).

INK WRITING

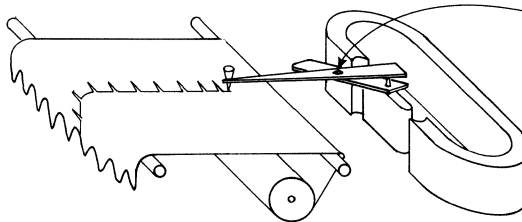


(A) CURVILINEAR RECORDER

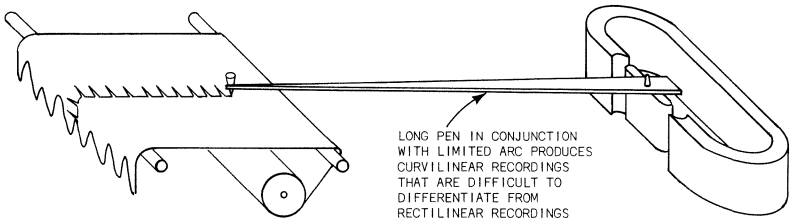
THERMAL WRITING



INK WRITING



INK WRITING PSEUDORECTILINEAR



(B) RECTILINEAR RECORDERS

Fig. 25-2. Curvilinear and rectilinear galvanometric recorder mechanisms.

25.1 BASIC RECORDER MECHANISMS

galvano-
metric
mechanisms

An input current or voltage is converted to mechanical movement of the stylus by either a galvanometric or a potentiometric principle. Galvanometric recorders utilize a moving coil and magnet assembly with a stylus arm attached to the moving coil. This assembly is somewhat similar to the common d'Arsonval galvanometer assembly used in most conventional panel meters; however, the assembly used in a graphic recorder must develop considerably more torque than the assembly used in a panel meter to overcome forces associated with stylus pressure and to provide good transient performance (damping). The various recorder types shown in Fig. 25-2 all use galvanometric recorder mechanisms. These mechanisms are normally constructed so that the angular deflection of the stylus arm is proportional to the magnitude of input current, that is, the relationship between input current and angular deflection is linear.

damping

Damping is particularly important in a galvanometric recorder. If the moving-coil mechanism is at rest and a current is suddenly applied to the coil, the coil, the stylus arm and stylus will commence to move to a new position. The stylus and arm will gain momentum during this movement and when the new equilibrium position is reached this momentum will cause them to overshoot this position. The process is then reversed with the stylus and arm oscillating about the equilibrium position for some time until finally coming to rest. Damping must be added to a galvanometric recorder mechanism to overcome these oscillations, the mechanism being referred to as critically damped when the stylus and arm assume an equilibrium position as quickly as possible without overshooting the position. It is most important that pen-to-paper friction constitute only a very small part of the pen system damping, which should be supplied electrically, otherwise the damping will be unreliable over a period of time, and will tend to vary from one side of the chart to the other.

mechanism
frequency
response

Due to finite force developed by a galvanometric recorder mechanism and the finite mass associated with the stylus and arm, the maximum angular acceleration of the system is defined and thus the high frequency response of the system is defined. Most galvanometric recorder mechanisms offer a maximum high frequency response of less than 200 Hz, the high frequency response being directly proportional to the force developed by the galvanometric mechanism and inversely proportional to the moment of inertia of the coil, stylus arm and stylus assembly.

potentiometric
mechanism

Potentiometric recorders operate on a servo principle with the position of the stylus arm being detected via a contact mechanism attached to the arm and in contact with a fixed slide-wire as shown in Fig. 25-3. The servo system will move the stylus arm and stylus until the potential between the slide-wire contact and the input voltage is zero. The linearity and accuracy of a potentiometric recorder mechanism is dependent only on the characteristics of the slide wire and associated electronics, that is, the relationship between input voltage and the position of the contact on the slide wire is linear. Potentiometric recorder mechanisms in general provide higher accuracy and better linearity than galvanometric mechanisms.

Damping in a potentiometric recorder mechanism is normally achieved electronically within the servo amplifier. Due to the inherent speed limitations of the servo system and the finite mass and frictional forces associated with the stylus arm, stylus and slide wire, potentiometric recorder mechanisms are inherently low frequency devices and rarely offer a high frequency response in excess of 20 Hz unless stylus deflection is limited to a centimeter or so.

galvanometric
versus
potentiometric

Since high accuracy and linearity is rarely of paramount importance in biophysical measurements and since a high frequency response of 20 Hz is not adequate for most biophysical recording applications, galvanometric recorders are generally preferred to potentiometric recorders for biophysical applications. Galvanometric recorders are also, in general, less expensive than potentiometric recorders, particularly if multi-channel instruments are considered.

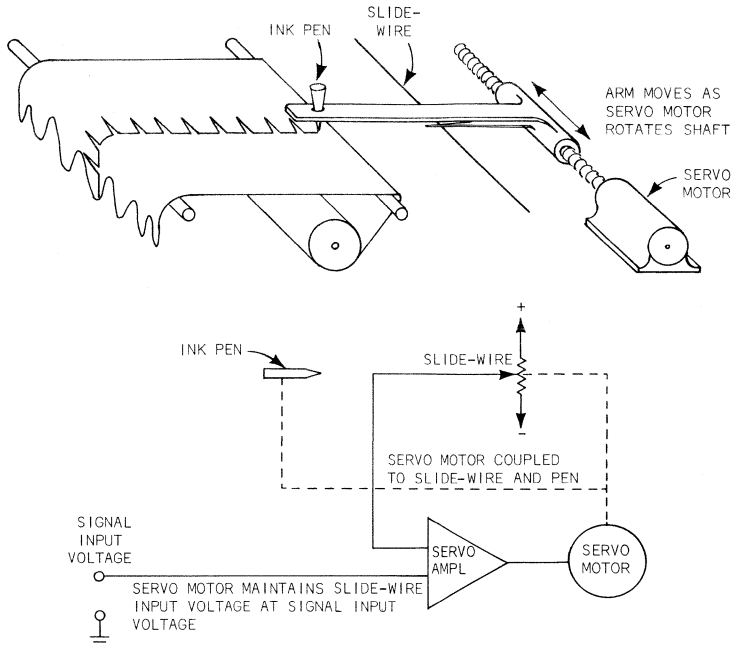


Fig. 25-3. Rectilinear, potentiometric (servo) recorder mechanisms.

25.2 RECORDING FORMATS

curvilinear

The geometrical relationship between stylus movement and the chart paper is either referred to as curvilinear or rectilinear. A pivoted stylus arm will transcribe an arc at its tip if the arm is caused to rotate about its pivot point. A stylus at the tip of the arm will, therefore, transcribe a curved line on stationary chart paper located beneath the stylus. Since the line is curved, the recording format is referred to as curvilinear recording. The basic galvanometric mechanism shown in Fig. 25-2A produces a curvilinear recording. A typical ECG and sinewave recorded with a curvilinear recorder is shown in Fig. 25-2A. Although the chart paper used with curvilinear recorders is marked with a curvilinear grid, the distortion produced by curvilinear recording is, at least, an inconvenience and thus rectilinear recorders are preferred. Curvilinear recorders are, however, considerably less expensive than rectilinear recorders and are extensively used, particularly for less critical applications such as may be encountered in a student laboratory.

rectilinear

Movement of a stylus in a straight line perpendicular to the direction of movement of the chart paper would transcribe a straight line on stationary chart paper located beneath the stylus. Since the line is straight, the recording format is referred to as rectilinear recording. The potentiometric mechanism shown in Fig. 25-3 produces a rectilinear recording. Rectilinear recording may also be achieved with the galvanometric recorder mechanisms shown in Fig. 25-2B.

geometry
causes
error

The thermal writing rectilinear recorder shown in Fig. 25-2B transcribes a curvilinear arc at the end of the stylus arm; however, the contact between the arm and the chart paper effectively moves up and down the stylus arm as the arm rotates about its pivot point, a straight line is, therefore, transcribed onto stationary chart paper. In analyzing the geometry of this technique it is apparent that the movement of the stylus on the chart paper is not directly proportional to the angular deflection of the stylus arm, thus producing a linearity error in the recorder. This linearity error is less than 1% if the total stylus deflection in both directions from its central resting point is limited to half the perpendicular distance between the "knife edge" and the pivot point.

Various recorder manufacturers have designed ingenious mechanical linkages between the stylus arm and chart paper to produce true rectilinear recording from a galvanometric mechanism as shown in the ink writing recorder in Fig. 25-2B. If stylus deflections are small when compared with the length of the stylus arm then the recording obtained, while being a true curvilinear recording, is difficult to differentiate from a rectilinear recording. This technique is extensively used in EEG recorders and is shown diagrammatically in the ink-writing, pseudo-rectilinear recorder in Fig. 25-2B.

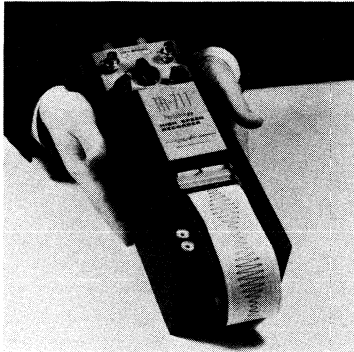
25.3 WRITING PRINCIPLES

The stylus in a graphic recorder must cause a written record to appear on the chart paper as the stylus moves across the surface of the paper. This written record may be achieved with a pen attached to the stylus arm, a heated stylus attached to the stylus arm, or with a rounded point attached to the stylus arm. The writing techniques are referred to as ink writing, thermal writing and pressure writing, respectively.

ink writing Ink-writing graphic recorders use ink pens operating on a capillary and siphon principle to record on untreated chart paper. The ink pens need constant attention to prevent blockage; however, the recording paper used is conventional paper, being considerably less expensive than the treated papers used for either thermal writing or pressure writing recorders. Thus, the ink-writing recorder is preferred in applications where a large volume of recording paper is required such as in electroencephalography.

thermal writing The thermal writing recorder uses a heated stylus in conjunction with specially coated chart paper. The stylus is heated by electric current flowing through a resistive element at the end of the stylus arm, as the heated stylus travels across the chart paper it melts off a white coating from the chart paper to expose a dark undersurface. While the thermal writing recorder requires little maintenance, the specially treated paper is somewhat expensive making this technique unsuited to applications requiring long term recording.

pressure writing Pressure writing recorders also utilize special chart paper. Carbon treated paper is transcribed by a rounded stylus at the end of the stylus arm with the pressure of the stylus leaving a black trace on the paper. Pressure writing recorders in general exhibit poor high speed characteristics due to the finite pressure required and, thus, the finite frictional forces involved at the stylus-paper interface.



TECHNI-RITE ELECTRONICS INC.
MODEL TR711

3dB FREQUENCY RESPONSE FROM DC TO 35Hz
FOR 4cm P-P DEFLECTIONS AND FROM DC TO
125Hz FOR 0.8cm P-P DEFLECTIONS.

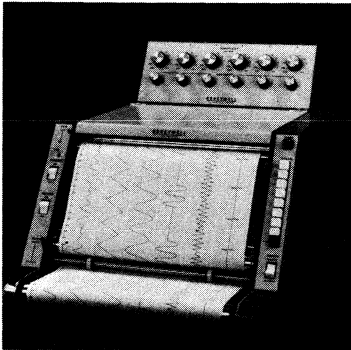
RECTILINEAR THERMAL WRITING

1mm/s TO 50mm/s

AC POWERED

WEIGHT - 10 POUNDS

A SINGLE-CHANNEL, PORTABLE, HIGH-SPEED RECORDER



HONEYWELL MODEL 2506 SHOWN WITH
MODEL 125 ATTENUATOR

FLAT FREQUENCY RESPONSE FROM DC TO 30Hz
FOR 4cm P-P DEFLECTIONS AND FROM DC TO
80Hz FOR 1cm P-P.

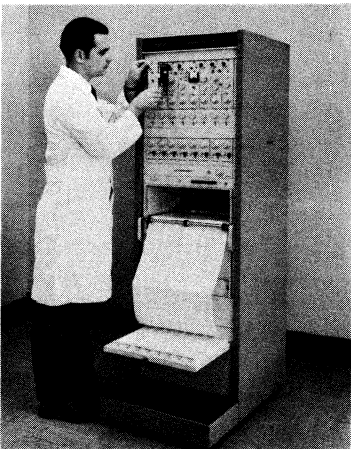
RECTILINEAR INK WRITING

1mm/min TO 250mm/s

AC POWERED

WEIGHT - 65 POUNDS

A SIX-CHANNEL HIGH-SPEED RECORDER



BECKMAN INSTRUMENTS INC.
TYPE S11 DYNOGRAPH *

FLAT FREQUENCY RESPONSE FROM DC TO 42Hz
FOR 4cm P-P DEFLECTION AND FLAT WITHIN
 $\pm 30\%$ FOR 0.8cm P-P DEFLECTION.

RECTILINEAR PRESSURIZED INK
WRITING

0.1mm/min TO 250mm/s

AC POWERED

RACK MOUNTED

*TRADEMARK BECKMAN INSTRUMENTS, INC.

AN EIGHT-CHANNEL HIGH-SPEED RECORDER

Fig. 25-4. Typical commercial recorders.

25.4 COMMERCIAL GALVANOMETRIC RECORDERS

Three commercial high speed recorders are shown in Fig. 25-4. All of these recorders are suited to biophysical measurements, providing chart speeds consistent with biophysical signals containing frequencies to 100 Hz or so. The single channel recorder shown in Fig. 25-4 is ideally suited for use in conjunction with the Tektronix Type 410 Physiological Monitor. The Type 410 provides a high level signal output via a rear panel connector which can be directly connected to the single channel recorder. Adequate output is provided by the Type 410 for use with almost any recorder and is directly compatible for use with any of the recorders shown in Fig. 25-4 as all of these recorders include input signal attenuation.

Type 410
recorder
output

25.5 SPECIAL PURPOSE RECORDERS

The preceding discussion has covered the recorder types in common use for biophysical measurements; however, many different types of graphic recorders, such as optical recorders and x - y recorders, are in common use and are occasionally used for biophysical measurements.

low cost
recorder

A low cost, single-channel, low speed recorder is shown in Fig. 25-5. This recorder is limited to low speed applications providing a maximum chart paper speed of 0.5 inches per second. This recorder uses a galvanometric movement that is almost identical to the movement used in most panel meters. The stylus arm is completely free to move and, thus, no frictional forces are involved. Once every second a mechanical lever forces this stylus arm and stylus against pressure-sensitive chart paper, leaving a small impression. Since the chart speed is limited, the written information on the chart paper appears as a series of closely spaced dots. This form of recorder is particularly suited for recording very low frequency physiological signals such as body temperature, heart rate variation, etc. Very slow chart speeds are available to allow 24 hours of information to be recorded on less than 2 inches of chart paper.



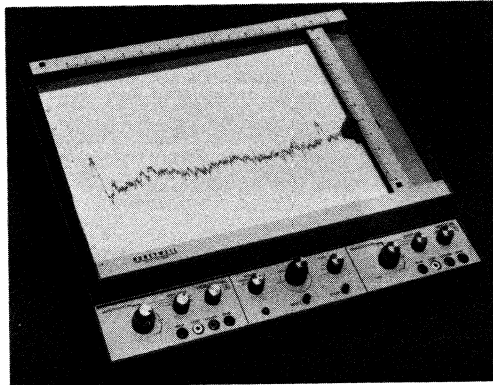
RUSTRAK INSTRUMENT CO.
MODEL 88

FOR DC OR LOW FREQUENCY USE ONLY
RESPONSE TIME IS $\approx 0.8s$

RECTILINEAR PRESSURE-SENSITIVE
PAPER WRITING SYSTEM
0.06in/h TO 0.5in/s

AC OR LOW VOLTAGE DC POWERED
WEIGHT - 5 POUNDS

Fig. 25-5. A single-channel low speed, low cost recorder.



HONEYWELL MODEL 560

11in x 16in PAGE SIZE
DC TO 5Hz (3dB)

INK WRITING

AC POWERED

TIME BASE
0.02 TO 5.0cm/sFig. 25-6. An $x-y$ recorder.

optical
recorder

Many graphic recorders use optical coupling between the basic recorder mechanism and the chart paper, the "chart paper" consisting of photographic film. High frequency performance to several kHz is obtainable with these optical recorders; however, in most instances, if high frequency performance in excess of that obtainable with conventional graphic recorders is desired, an oscilloscope and continuous motion camera is preferable to an optical recorder for they are far less susceptible to mechanical shock and vibration and, in general, far more versatile.

$x-y$ recorder

An $x-y$ recorder is a device used to plot two variables against one another on chart paper rather than, as with a conventional graphic recorder, plotting one variable against time on chart paper. A single sheet of graph is commonly used as the chart paper, the stylus moving in both x and y directions across the paper. A typical $x-y$ recorder is shown in Fig. 25-6. The transient response performance of $x-y$ recorders limits them to low frequency applications. Except where extreme accuracy is required, an oscilloscope and conventional oscilloscope camera is normally preferred to the $x-y$ recorder as the combination offers superior transient performance and greater versatility.

26

MAGNETIC TAPE RECORDERS

A magnetic tape recorder is an analog storage device. *Electrical* signals fed into a magnetic tape recorder are stored on the magnetic tape with the same *electrical* signals available from the magnetic tape recorder when required. Although the oscilloscope camera and the graphic recorder are information storage devices, once information has been recorded by these devices it is extremely difficult to regenerate the information in its original electrical format. The magnetic tape recorder is the only convenient device that permits recording of data in such a manner as to allow it to be reproduced at some later time in its original electrical format, thus, making it a "holding" device as well as a recording device. A magnetic tape recording may be referred to as a "soft copy" as it is capable of being reconverted to the original electrical variable. The oscilloscope camera and graphic recorder produce "hard copy" which is not capable of convenient reversion.

"holding"
device

soft/hard
copy

The magnetic tape recorder allows biophysical signals to be recorded in real time during a physiological experiment and then allows the experimental results to be reproduced at some later time. This allows those performing the experiment to concentrate on obtaining data during the experiment and then to concentrate on analyzing the data after the experiment has been completed. Some monitoring of the recorded signal must be used, however, as it is frequently too late to correct some obvious fault in the recording system unless signals are monitored while they are being stored on magnetic tape.

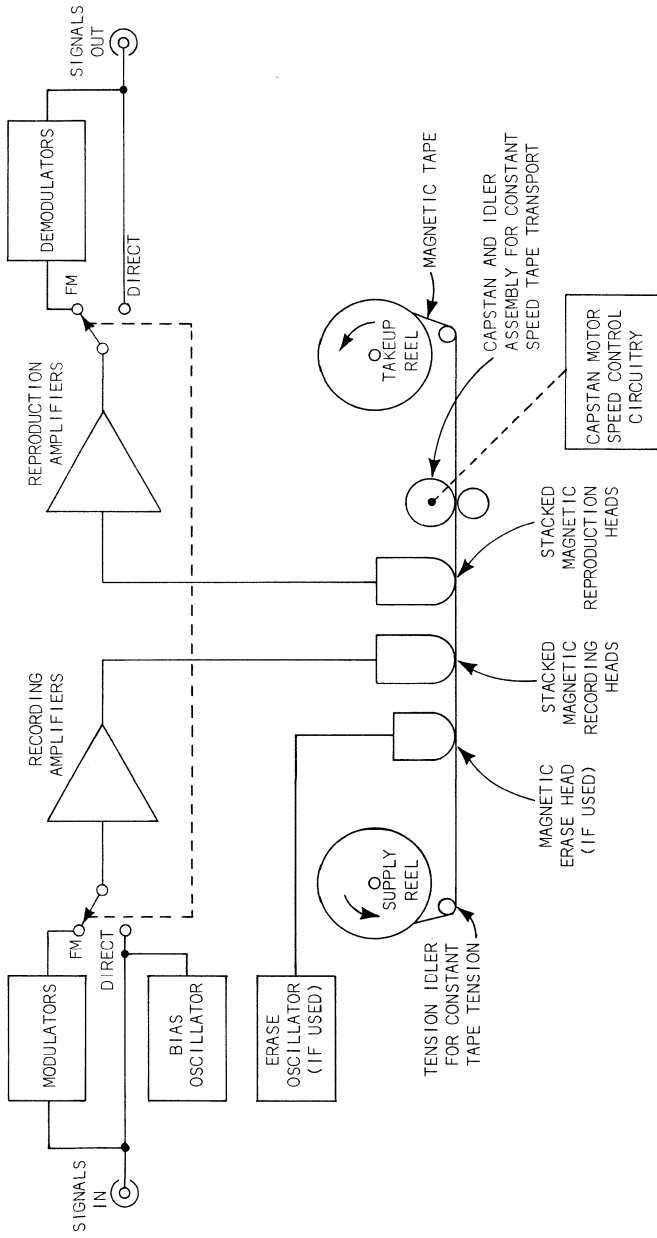


Fig. 26-1. Basic components of an instrumentation tape recorder.

record slow,
 playback
 fast

record fast,
 playback
 slow

Since most magnetic tape recorders have the capability of recording information at one speed and then reproducing it at a different speed, time expansion or compression of the information can be accomplished. By recording at a slow speed and reproducing at a faster speed, many hours of experimental data can be reproduced in just a few minutes, thus taking full advantage of the capabilities of other instruments in the measuring system such as the high speed capability of data processing equipment or the high frequency capability of a spectrum analyzer. By recording information at a high speed and reproducing it at a slower speed, high frequency information is reproduced as low frequency information, thus allowing more time for detailed analysis of the information and allowing the subsequent use of low frequency instrumentation such as a graphic recorder.

Two basic recording techniques are used in instrumentation magnetic tape recorders suitable for biophysical applications -- direct recording and indirect frequency-modulation (FM) recording. These two techniques are covered separately in Sections 26.1 and 26.2.

26.1 DIRECT MAGNETIC TAPE RECORDING

The following discussion assumes that the reader is familiar with magnetic tape recorder principles as they relate to the unsophisticated audio tape recorder commonly used for home entertainment. These recorders all use the direct recording technique.

Fig. 26-1 shows the basic components of an instrumentation tape recorder and shows switching in both the recording circuitry and the reproduction circuitry for switching between DIRECT or FM. The tape recorder is operating in the direct recording mode with the switch in DIRECT. Referring to Fig. 26-1, in the direct mode the input signal is algebraically added to a high frequency bias signal of about 100 kHz and then fed directly into a recording amplifier which provides the recording current necessary to drive the magnetic recording head. The recording head converts the current into a varying magnetic flux which changes the residual magnetism on the magnetic tape as it moves past the recording head. This magnetism on the tape, in turn,

recording-
 circuit
 description

produces a change in the output voltage from the magnetic reproduction head. The voltage produced at the reproduction head is then amplified in a reproduction amplifier to provide a high level output signal.

Direct magnetic tape recording has two severe disadvantages -- poor low frequency response and poor output level stability. The poor low frequency response is due to the magnetic characteristics of the recording and reproduction heads as the efficiency of the heads decreases as frequency drops below a few hundred hertz. For biophysical applications it can be stated that direct magnetic tape recording is unsatisfactory for signals below 100 Hz.

"direct"
mode
limitations

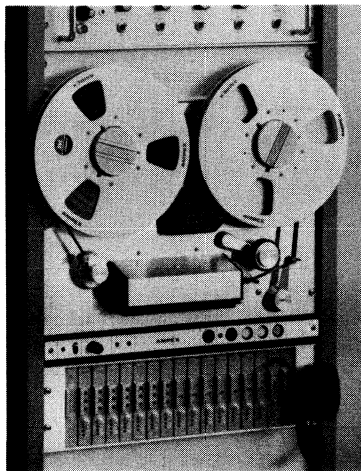
If a constant-level, audio-frequency signal is recorded on a home entertainment audio tape recorder, the level of the output signal will fluctuate by 30% or more primarily due to variations in the magnetic coating on the tape and, also, to variations encountered in the contact between the tape and the recording and reproduction heads. This 30% variation, while being quite acceptable for home entertainment recorders, is unsatisfactory for biophysical measurements. By the use of high quality magnetic tape and careful design of the tape transport system, this amplitude stability can be held within 10% or so. This 10% output level variation exists on most instrumentation tape recorders operating in the direct mode and is a severe limitation in many applications. This limitation is completely overcome by recording via the indirect FM technique and this technique also extends low frequency response to DC.

26.2 INDIRECT FM MAGNETIC TAPE RECORDING

Referring to Fig. 26-1 with the recorder operating in the FM mode as shown, the input signal is applied to a modulator where it is used to frequency modulate a carrier frequency, this modulated carrier is then fed to the recording amplifier which provides the recording current necessary to drive the magnetic recording head. The subsequent voltage from the reproduction head is then amplified and demodulated to provide an output signal. Indirect FM tape recording permits frequencies down to and including DC to be recorded.

"FM" mode
limitations

The high frequency response of any magnetic tape system is limited by the characteristics of the recording and reproduction heads, by the signal-to-noise ratio that is acceptable, and to some extent by the characteristics of the magnetic recording tape used. Since the high frequency response of the magnetic tape system is limited and thereby limiting the FM carrier frequency, the high frequency response of the indirect FM system is further limited due to the maximum deviation limits of the FM modulator and demodulator (normally about 20% of the FM carrier frequency).



DIRECT OR FM RECORD/REPRODUCE SYSTEMS
 SIX TAPE TRANSPORT SPEEDS
 FROM 15/16in/s TO 30in/s
 USES EITHER 1/4in WIDE OR 1/2in WIDE
 MAGNETIC TAPE
 SEVEN DATA CHANNELS AND AN ADDITIONAL
 VOICE CHANNEL FOR VOICE ANNOTATIONS
 DIRECT RECORD/REPRODUCE SYSTEM:
 100Hz TO 150kHz AT 30in/s
 50Hz TO 5kHz AT 15/16in/s
 FM RECORD/REPRODUCE SYSTEM:
 DC TO 10kHz AT 30in/s
 DC TO 312Hz AT 15/16in/s
 PORTABLE CASE OR RACK MOUNTED
 WEIGHT - 80 POUNDS

Fig. 26-2. The Model PR - 500 seven-channel instrumentation recorder (Ampex Corporation).

sidebands
 limit
 bandwidth

Referring to the instrumentation recorder shown in Fig. 26-2, the specifications for the high frequency response of the direct record/reproduce system is only 5 kHz at a tape speed of 15/16 inches per second. The maximum frequency component in the modulated FM carrier frequency must, therefore, be below 5 kHz. Since 8 sidebands are required on each side of an FM carrier frequency for almost distortion free FM reproduction, the high frequency response of the recorder in the FM record/reproduce mode is, thus, limited to *one-sixteenth of the high frequency response* of the direct record/reproduce system or only 312 Hz at a tape speed of 15/16 inches per second. As the tape speed of the system is increased, the high frequency performance of the recording and reproduction heads is increased, the FM modulation frequency can be increased, and, therefore, the high frequency performance in the FM record/reproduce mode is proportionally increased.

In the indirect FM recording mode, amplitude stability between the signal applied to the recording head and the signal received from the reproduction head need not be considered because all instability will either be removed in a limiting stage in the reproduction amplifier or will be of no consequence due to the insensitivity of the demodulator to amplitude modulated signals.

26.3 TAPE TRANSPORT MECHANISM

The basic components of the tape transport mechanism used in an instrumentation tape recorder is shown in Fig. 26-1. The tape supply reel may supply either 1/4- or 1/2-inch wide tape from a 10 1/2-inch diameter reel which supplies 3600 feet of normal thickness tape. A rotating capstan and idler assembly pull this tape past the magnetic heads at a constant speed; the tape is then collected on a takeup reel. Tension idler assemblies on both sides of the capstan are used to provide constant tension on the tape.

- tape speeds The speed at which the magnetic tape is drawn past the magnetic heads is controlled by the speed of the capstan motor which drives the capstan and idler assembly. Most instrumentation tape recorders offer a speed range from 15/16 in/s to 30 in/s with four intermediate speeds. Some instrumentation recorders operate up to speeds of 240 in/s.
- tape speed stability Tape speed stability is an important consideration in instrumentation recorders. Relative tape speed instability is referred to as "wow" if occurring at less than about 2 Hz and "flutter" if occurring at greater than about 2 Hz. With the recorder operating in the direct mode, any instability in the tape speed will appear as a change in frequency of the output signal. With the recorder operating in the FM mode, any instability in the tape speed will appear as an amplitude variation in the output signal. Control circuitry within instrumentation tape recorders maintains a constant tape speed by using some form of servo mechanism; the feedback to the servo is derived via a tachometer attached to the capstan or via the output from a constant frequency source that has been recorded onto one channel of a multi-channel recorder.

26.4 OTHER MAGNETIC TAPE RECORDER CONSIDERATIONS

erasure If a magnetic tape has been used previously to record information, this information must be erased from the tape before the tape can be reused for recording. Erasure can be achieved with a bulk tape eraser that exposes a complete reel of tape to an intense low frequency magnetic field. Erasure may also be accomplished via a magnetic erase head on the tape recorder; this head exposes the tape to an intense high frequency magnetic field before the tape reaches the magnetic recording head. Many instrumentation tape recorders, particularly multi-channel recorders, do not incorporate erasure as it is generally more convenient and less expensive to use a bulk eraser.

track width The previous discussion assumes only one channel of information being recorded on the tape recorder. Most commercial instrumentation tape recorders offer more than one channel of information; the recorder shown in Fig. 26-2 offers seven channels for information plus an additional channel for voice annotations. Obviously, if seven tracks plus a voice track are to be stacked onto a 1/4-inch wide tape, the width of each track must be extremely narrow. The instrumentation recorder shown in Fig. 26-2 uses track widths of 0.024 inch.

The tape transport used in an instrumentation tape recorder must be capable of quickly winding tape in the forward direction to allow a desired segment of tape to be reached quickly. Similarly, the transport must be capable of quickly rewinding the tape back onto the supply reel. The braking system incorporated into the tape transport mechanism must be capable of quickly stopping tape movement when a stop command is received.

noise

The maximum achievable signal to noise ratio in the best quality tape recorders, in either the direct or FM modes, is about 44 dB, it is thus necessary to record at as high a signal level as the system will permit and to use some filtering during playback to reduce the noise as much as possible. The signal to noise ratio in the FM mode may be expected to approach 44 dB, however in the direct mode the signal to noise ratio will rarely exceed 30 dB.

Instrumentation grade magnetic tape should be used for physiological recording, in contrast to lower quality entertainment grade tape. The use of instrumentation grade tape will insure that the maximum signal to noise ratio of the system is obtained, it will insure that the maximum amplitude stability in the direct mode is obtained and it will insure that there is no "information dropout" due to "dropouts" on the tape. Careful maintenance is particularly important with magnetic tape recorders as the signal to noise ratio of the recorder will be drastically reduced should the record and/or playback heads become worn, dirty and/or magnetized.

Manufacturers maintenance instructions usually include head cleaning and demagnetizing procedures.

27

DATA TRANSMISSION AND PROCESSING

Transmission of data between various elements in a biophysical measuring system is normally accomplished directly with shielded cable. Shielded cable is, however, unsuited for many data transmission requirements, particularly for transmission over long distances or if it is desired for one item of instrumentation to be completely isolated both physically and electrically from other items in the system. Data transmission via a radio link, referred to as telemetry, is used in these applications. Data transmission becomes particularly important if the physiological signal is to undergo data processing in a computer as, in most instances, the computer is located remotely from the data source.

27.1 DATA TRANSMISSION VIA SHIELDED CABLE

Data transmission via shielded cable is the simplest form of data transmission; the signal is transmitted in its original analog voltage form. For faithful transmission, the transmission medium must be capable of passing the physiological signal over its maximum conceivable amplitude and bandwidth range. In a direct-wire unterminated transmission link using shielded cable, losses in signal amplitude are caused by cable reactance and are in direct proportion to cable length. For most physiological data transmission applications, cable reactance imposes a practical maximum cable length of about 1,000 feet when the cable is driven by normal instrumentation providing signals from an output impedance of perhaps 100 ohms.

Within the bandwidth of interest in biophysical measurements (DC to 30,000 Hz) the capacitance from the shielded cable inner conductor to the shield is the most important single factor limiting the high frequency performance of the transmission system. The bandwidth of a shielded cable transmission system, when driven from a voltage source, is given by:

$$f_{3dB} = \frac{.16 \times 10^{12}}{RC\ell}$$

determining
cable system
bandwidth

R is the output resistance of the driving source in ohms, C is the nominal capacitance of the shielded cable in picofarads per foot and ℓ is the length of the cable in feet. Instrumentation providing an output signal via a vacuum-tube cathode-follower circuit provides a typical driving source impedance of $\leq 1,000$ ohms; instrumentation providing an output signal via a transistor emitter-follower circuit provides a typical driving source impedance of ≤ 100 ohms. Most shielded cable has a capacitance of between 20 and 40 pF/ft; the common 1/4-inch diameter shielded cable used in many physiological laboratories exhibits a typical capacitance of 30 pF/ft. Relating this information to the previous formula, a 1,000-foot length of 30 pF/ft shielded cable, when driven from a source impedance of 100 ohms, provides a transmission system bandwidth of 50 kHz. Practically, the system bandwidth will be somewhat less than this as the driving amplifier does not "see" only the capacitance of the line, but its inductance and resistance as well. At 50 kHz, a 1,000-foot line, allowing for propagation velocity in the cable, is a significant fraction of a wavelength long.

Although unterminated direct wire transmission links using shielded cable can be used to about 1,000 feet it is recommended that terminated systems be used for analog data transmission in excess of 50 feet. Terminated systems require the

shielded cable to be driven from a source of its own characteristic impedance and be terminated in its characteristic impedance. To avoid ground loops, it is customary, and usually essential, to transmit from a single-ended driver having the line characteristic impedance, and to receive by a differential amplifier between line and shield having a high common mode impedance, and a differential input impedance equal to the characteristic impedance of the line.

The overall bandwidth of a biophysical measurement system is given by:

$$f = \frac{1}{\sqrt{\frac{1}{f_1^2} + \frac{1}{f_2^2} + \frac{1}{f_3^2} + \dots \text{etc.}}}$$

where f_1 , f_2 , f_3 , etc. are the bandwidths (upper -3 dB frequency limits) of the individual components in the system including the transmission components between individual instruments.

AM techniques

Wire transmission can be used over very long distances by employing an amplitude modulated carrier frequency with the information to be transmitted being used to amplitude modulate a carrier frequency. This carrier frequency must be higher than the highest frequency component in the modulating signal by a factor of at least five to allow low-distortion demodulation at the receiver. Any amplitude modulated transmission system is only as good as the ability of the system to faithfully reproduce amplitude variations. System attenuation, system noise, system nonlinearity and/or interference from other sources produce errors in the information transmitted. For this reason, amplitude modulated data transmission techniques are limited to wire transmission rather than to radio link transmission.

27.2 DATA TRANSMISSION VIA A TELEMETRY LINK

Telemetry links used in biophysical measurements range from the short range systems used in behavioral studies laboratories to the extremely long range systems used in the aerospace industry. The principal difference between the short range and long range systems is the power output capability of the transmitter and the sensitivity of the receiver.

short
range
applications

Short range telemetry systems are used in behavioral studies to completely isolate a subject or laboratory animal from a recording system. A small telemetry transmitter may be attached to the subject to transmit, for example, the subject's ECG; a telemetry receiver may be located only a few feet away to allow this ECG signal to be processed by a physiological measurement system. This allows complete subject freedom of movement and thus allows for more natural subject behavior. As a telemetry transmitter is normally a completely isolated battery operated device, telemetry offers a degree of safety unattainable with conventional instrumentation due to the complete elimination of direct electrical connections between the subject and the instrumentation.

single-
channel
telemetry

Short range telemetry systems usually use FM techniques and may transmit in the FM broadcast band (88 to 108 MHz). They may be simple systems providing only one data channel or more complex systems providing multichannel capability by using subcarrier modulation. A typical single-channel telemetry link is shown diagrammatically in Fig. 27-1. The single-channel battery-powered FM transmitter shown would normally be one cubic inch or so in volume providing several hundred hours of operation from one or two miniature mercury batteries. The receiver must provide DC coupling from the discriminator to give the system a DC signal transmission capability, and, to avoid DC drift, both transmitter and receiver should be crystal controlled.

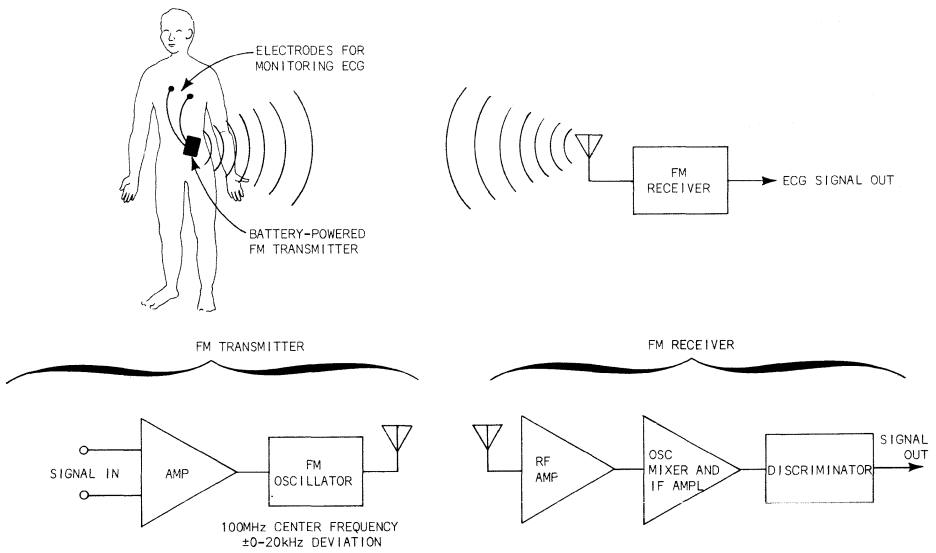
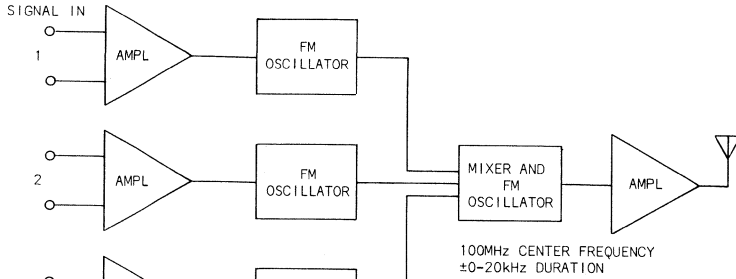


Fig. 27-1. A typical single-channel telemetry link.

FM/FM TRANSMITTER



SUBCARRIER OSCILLATORS

- 1 3kHz CENTER FREQUENCY
±0-225Hz DEVIATION
- 2 3.9kHz CENTER FREQUENCY
±0-293Hz DEVIATION
- 3 5.4kHz COMPUTER FREQUENCY
±0-405Hz DEVIATION

IRIG PROPORTIONAL SUBCARRIER BANDS (Hz)

BAND	LOWER LIMIT	CENTER FREQ	UPPER LIMIT	BAND WIDTH	NOMINAL INTELL. FREQ
1	370	400	430	60	6
2	518	560	602	84	8
3	675	730	785	110	11
4	888	960	1,032	144	14
5	1,202	1,300	1,398	196	20
6	1,572	1,700	1,828	256	25
7	2,127	2,300	2,473	346	35
8	2,775	3,000	3,225	450	45
9	3,607	3,900	4,193	586	60
10	4,925	5,400	5,805	810	80
11	6,799	7,350	7,901	1,102	110
12	9,712	10,500	11,288	1,576	160
13	13,412	14,500	15,588	2,176	220
14	20,350	22,000	23,650	3,300	330
15	27,750	30,000	32,250	4,500	450
16	37,000	40,000	43,000	6,000	600
17	48,560	52,500	56,440	7,880	790
18	64,750	70,000	75,250	10,500	1,050
19	86,025	93,000	99,975	13,950	1,400
20	114,700	124,000	133,300	18,600	1,900
21	152,625	165,000	177,375	24,750	2,500
A	18,700	22,000	25,300	6,600	660
B	25,500	30,000	34,500	9,000	900
C	34,000	40,000	46,000	12,000	1,200
D	44,620	52,500	60,380	15,760	1,600
E	59,500	70,000	80,500	21,000	2,100
F	79,050	93,000	108,950	27,900	2,800
G	105,400	124,000	142,600	37,200	3,700
H	140,250	165,000	189,750	49,500	5,000

FM/FM RECEIVER

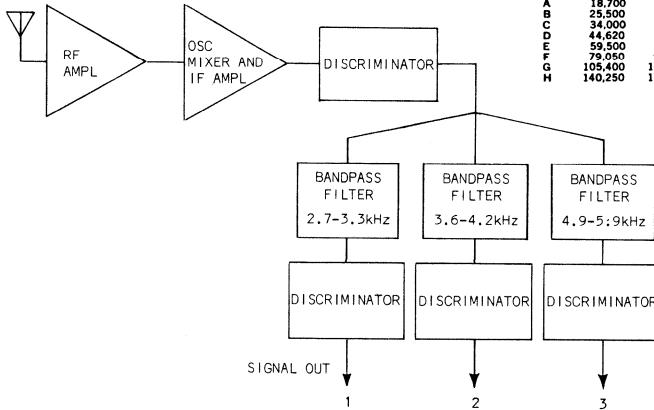
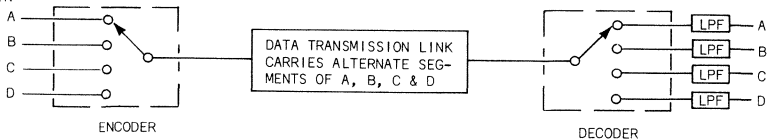


Fig. 27-2. A typical multichannel telemetry link — frequency division multiplexed.

PHYSIO-LOGICAL SIGNALS IN



ENCODER AND DECODER KEPT IN SYNC BY THE ADDITION OF A SYNCHRONIZING SIGNAL IN THE ENCODED INFORMATION

LPF - LOW PASS FILTERS TO REMOVE MULTIPLEXING-FREQUENCY COMPONENT IN THE OUTPUT SIGNALS

Fig 27-3 Principle of time division multiplexing

FM/FM system

A typical multichannel telemetry link is shown diagrammatically in Fig. 27-2. The input signal is used to frequency modulate subcarrier oscillators operating below 100 kHz; the frequency modulated outputs from these three subcarrier oscillators are mixed and used to frequency modulate a high frequency oscillator. This system is referred to as an FM/FM system as both the subcarrier oscillators and the main oscillator are frequency modulated. A list of IRIG FM/FM subcarrier frequency bands is shown in Fig. 27-2. The maximum signal that can be transmitted via the telemetry link is known as the nominal intelligence frequency and is directly proportional to the bandwidth of the subcarrier channel used. It can be seen from Fig. 27-2 that, at a subchannel bandwidth of 24.75 kHz, the nominal intelligence frequency is only 2.5 kHz. Many multichannel telemetry systems use wider channel bandwidths and thus require correspondingly greater channel separations. Subcarrier bands 1 through 21 shown in Fig. 27-2 have relatively narrow bandwidths; however, subcarrier bands A through H have wider bandwidths to allow higher frequency information to be transmitted.

27.3 TIME DIVISION MULTIPLEXING

frequency
division
versus
time
division

Multiplexing refers to the combining of several channels of information to allow this information to be transmitted via one data transmission link. The FM/FM telemetry system discussed previously is an example of frequency division multiplexing as the signals are all transmitted simultaneously using various *frequency* bands to separate the signals. Whereas frequency division multiplexing is true *simultaneous transmission* of separate channels of data, time division multiplexing consists of *sequential transmission* of separate channels of data. This multiplexing technique is best illustrated by the operation of a rotating commutator in which the several input signals are switched sequentially onto a common output channel. The multiplexed output may be transmitted directly over wire or used to modulate a high frequency carrier. At the receiving end the signal is separated back into individual channels by a decoder that is synchronized with the transmitting encoder. This technique is illustrated in Fig. 27-3. The frequency at which the encoder switches between channels should be substantially higher than the maximum information-frequency content of either of the channels.

other data transmission systems

Although FM/FM multiplexing and time division multiplexing are commonly used for biophysical applications, many other information transmission systems are used for special applications. These transmission systems are not unique to the biophysical sciences and any text on radio communications should provide a good reference as to other possible data transmission systems.

27.4 DATA PROCESSING WITH AN ANALOG COMPUTER

principle

The analog computer is a highly versatile signal processor. Analog computers consist of precision electronic modules such as operational amplifiers, voltage sources, voltage dividers, multipliers and other special devices. The analog computer is able to analyze nonelectronic systems by representing these systems in electronic units. Basic system quantities are normally represented by varying voltage or current; system constants are represented by either discrete resistors or by voltage dividers. Once a physical quantity has been converted to an electronic analog, the analog computer processes these electronic quantities by using various operational amplifier configurations for amplification, attenuation, integration, summation and numerous other functions.

A sample problem and its solution via an analog computer are shown in Fig. 27-4. The equation shown (Fig. 27-4) represents the behavior of a blood flow system when subjected to a step change in pressure. If this step change in pressure (p) is represented by a voltage step and the desired pressure in some other part of the flow system (y) is represented as an output voltage from the analog computer, then the differential equation relating p to y can be solved by the circuit shown. This circuit only uses five operational amplifiers and two precision attenuators; however, typical problems solved by an analog computer may involve the use of many more components. Commercial analog computers may vary in size from units offering only six operational amplifiers to extremely large installations offering several hundred operational amplifiers. The mechanical arrangement of operational amplifiers and other modules in analog computers is normally arranged so that interconnection between various modules can be achieved by a patching network.

PROBLEM - A PHYSIOLOGICAL BLOOD-FLOW ANALYSIS

A STEP PRESSURE CHANGE TO A FLOW SYSTEM (p) CAUSES A DAMPED SINUSOIDAL PRESSURE CHANGE (y) TO BE PROPAGATED THROUGHOUT THE SYSTEM. SYSTEM ANALYSIS YIELDS THE FOLLOWING SECOND ORDER DIFFERENTIAL EQUATION WITH SYSTEM CONSTANTS REPRESENTED BY $C1$, $C2$ AND $C3$.

$$C2 \frac{d^2 y}{dt^2} + C1 \frac{dy}{dt} + C3 y = p$$

SOLUTION - SOLVE THE ABOVE EQUATION WITH OPERATIONAL AMPLIFIERS AND CALIBRATED DIVIDERS.

SINCE THE EQUATION REPRESENTS A DYNAMIC SYSTEM, SOLUTION OF THE ABOVE WOULD YIELD A COMPLEX SERIES OF TERMS TO REPRESENT THE OUTPUT OF THE SYSTEM. THE ANALYSIS CAN BEST BE ACCOMPLISHED WITH AN ANALOG COMPUTER. THE ABOVE EQUATION CAN BE REWRITTEN:

$$\frac{p}{C2} - \frac{C1}{C2} \frac{dy}{dt} - \frac{C3}{C2} y = \frac{d^2 y}{dt^2}$$

SOLUTION POINT
LEFT SIDE = RIGHT SIDE

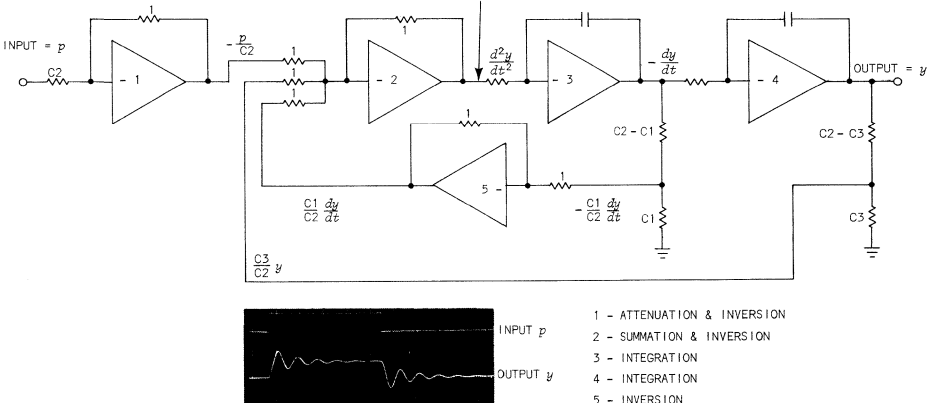


Fig. 27-4. Physiological system analysis with an analog computer.

27.5 DATA PROCESSING WITH A DIGITAL COMPUTER

time
sharing

With the advent of digital computer time sharing techniques, the use of the digital computer is now economically feasible for many biophysical measurement applications. While the central processing unit of a digital computer can normally only process information from one source at any one time, it can often process this information in a fraction of a second and then begin processing information from another source. This is referred to as time sharing. In a time shared system, due to the speed of the computer, the user is often unaware that the central processing unit is sharing its time between him and many other users.

digital-
analog
conversion

Data must be presented to, and is received from, a digital computer in a digital format. If analog data is to be processed with a digital computer, it must, therefore, be converted via an analog to digital converter. Conversely, if the data received from the digital computer is required in an analog format, it must be converted to this format in a digital to analog converter. If a computer is located remotely from the analog data gathering site, it is preferable to transmit and receive data in analog form. If the data is digitized on site, it must be either transmitted in parallel to the computer (one cable per bit), which is very expensive, or serially, which is also expensive and slow as well. It is quite feasible for ECG or EEG data, but impractical for direct brain recording or action potentials.

real time
processing

Since the time scale of analog data is usually important, this time scale must be maintained when processing analog data via a digital computer. It is thus necessary that the computer process this data in real time as shown in Fig. 27-5A. This precludes the direct use of a time shared computing system unless the priorities in the time shared system are arranged so that processing of this analog information takes absolute priority over any other processing. In this case, the time shared computer system would process this data in real time. For this reason, many self contained small computers are used in this application.

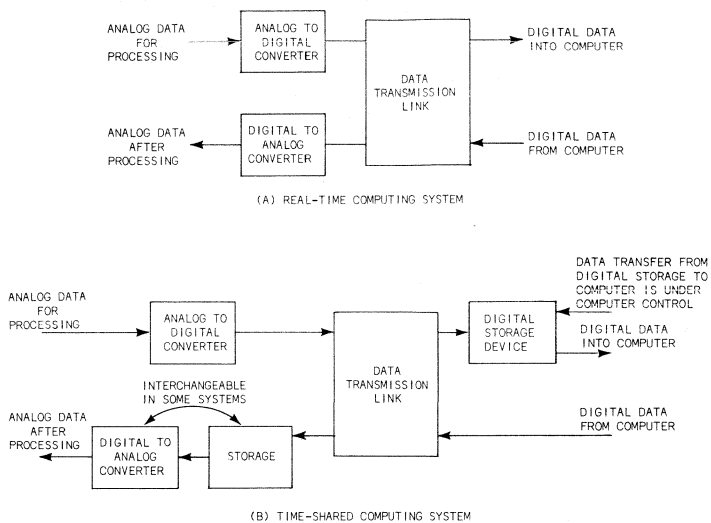


Fig. 27-5. Analog data to and from a digital computer.

Although the time scale of analog data is usually important and must be maintained during a data processing procedure, it is possible for a time shared computer system to indirectly process this data in other than real time and then to reconstruct the data in real time after a finite delay through the use of storage devices. Referring to Fig. 27-5B, analog data is converted to digital data via an analog to digital converter; this digital data is then stored in a digital storage device (normally located at the computer site). The time shared computing system then accepts data from the digital storage device on a time shared basis with transfer being controlled by the computer. Digital data from the computer will then be produced in segments as the computing system finds time to process stored data. These segments of digital data can then be converted to analog data and stored or they may be stored in a digital storage device before converting. All of the analog data can then be withdrawn from the storage device to preserve the time scale of the data.

preserving
time scale

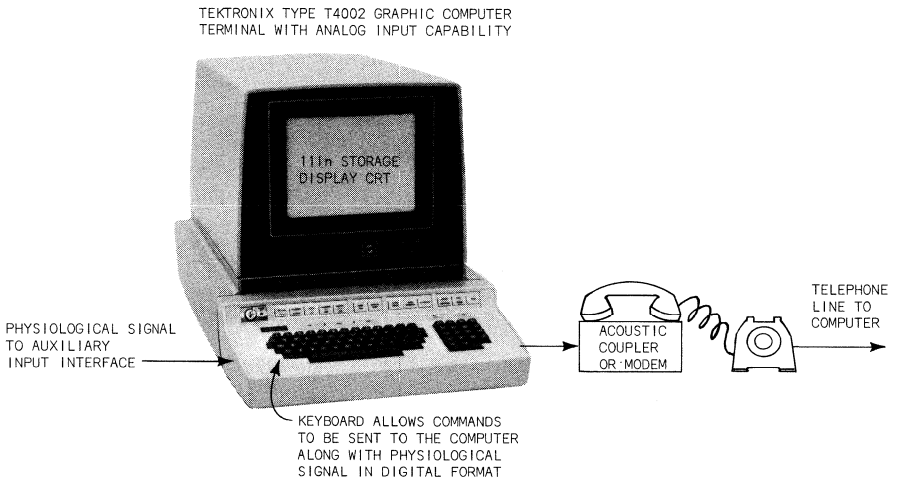


Fig. 27-6. A computer terminal used for physiological signal analysis and processing via a remote computer.

Type T4002

The Tektronix Type T4002 Graphic Computer Terminal shown in Fig. 27-6, when interfaced with a custom analog input module, is ideally suited to analog input data processing with a time shared computing system. The terminal, when used in conjunction with an acoustic coupler or "modem" (MODulator-DEModulator) and data communications service, provides the necessary electronics to implement the time shared computing system discussed previously and shown diagrammatically in Fig. 27-5. The storage function is achieved by using the storage cathode-ray tube of the terminal. The communication line provided by the data communication service should have sufficient bandwidth (baud rate capability) to allow data transmission in real time. The ECG signal typically requires a 400 baud line.

27.6 DATA PROCESSING APPLICATIONS AND SOFTWARE

Data processing applications in biophysical measurements require application software (computer programs) to control the computer in performing specific tasks. This software is continuously being developed by computer manufacturers, computer users and consulting firms and most computer manufacturers have a "library" of software programs available to their customers.

signal
averaging

The digital computer is particularly suited to signal averaging. Digital Equipment Corporation has a signal averaging program for their PDP-8/I computer that also permits prestimulus averaging and computation of statistical information such as standard deviation and trends. The Digital Equipment Corporation's PDP-8/I computer may be used in conjunction with the Tektronix Type 611 Storage Display Unit and an analog to digital converter as a complete signal averaging instrument.

ECG
analysis

Computers are now extensively used for analysis of ECG waveforms. The information storage and waveform comparison capability of the computer allow it to compare an ECG signal with other ECG signals previously analyzed to determine if the signal is within acceptable statistical limits. The Control Data Corporation provides software for electrocardiogram analysis to use with their Model 1704 Computer; the complete package of computer software and peripheral equipment is designated the "1700 Computer Electrocardiogram Analysis System." This system identifies, measures and analyzes waveforms of 12-lead electrocardiograms by performing a pattern recognition analysis on the electrocardiogram and provides a printout of the results for diagnostic use.

Computers are ideally suited to simulation of physiological systems, thus, in many cases, allowing an analysis of physiological systems within a computer rather than by performing actual physiological measurements.

INTENSIVE CARE CONCEPTS

intensive
care units

Most diseases of the heart and of the circulatory system, referred to as cardiovascular diseases, strike without warning and prompt treatment is required if death is to be averted. Such treatment is best provided in a specialized area of a hospital referred to as an "intensive care unit." These specialized hospital units provide constant observation of the subject, constant monitoring of the subject's physiological condition and provide immediate emergency treatment should it be required. Coronary intensive care units are used for treatment of diseases of the heart such as the myocardial infarction or "heart attack." Stroke intensive care units are used for treatment of diseases of the circulatory system such as the stroke. Pulmonary intensive care units are used for treatment of respiratory diseases.

An intensive care unit may consist of one or more subject-monitoring sites, referred to as "beds" as each site is, in fact, a bed. Electronic instrumentation at each subject-monitoring site monitors various physiological signals from one subject and may activate alarms should these physiological signals be above or below predetermined limits. The complete intensive care unit consists of not only the necessary monitoring equipment but also the necessary trained personnel and emergency equipment to allow immediate treatment of cardiac malfunctions.

Intensive care units are usually constructed to suit a particular hospital's requirements. Widely varying approaches are taken as to the physiological functions to be monitored and the type of monitoring equipment required and, in general, no two intensive care units are alike. Intensive care instrumentation is continuously being developed to accomplish more advanced physiological monitoring techniques. The following discussion on intensive care concepts is intended to present some of the physiological functions that *may* be monitored and some of the typical instrumentation that *may* be used to monitor these functions. The discussion is conceptual in nature and in no way attempts to survey all current intensive care applications and instrumentation.

28.1 PHYSIOLOGICAL FUNCTIONS TO BE MONITORED DURING INTENSIVE CARE

Since subjects in coronary intensive care units are suffering from cardiovascular diseases, all physiological functions associated with the heart and the circulatory system should be monitored. In monitoring these physiological functions, it should be borne in mind that the subject is in a recovery situation, thus, the physiological monitoring devices should not hinder the recovery process and should be as unobjectionable as possible to the subject.

ECG
monitoring

The principal physiological signal monitored in an intensive care unit is often the electrocardiogram. The electrocardiogram is usually monitored in the lead II configuration with two active electrodes placed approximately 12 inches apart along the maximum potential axis of the subject's heart. A third electrode (ground) should be located elsewhere on the chest. This electrocardiogram monitoring configuration is referred to as a three-lead chest cluster. Tektronix produces a patient cable for use with their Type 410 Physiological Monitor that is specifically intended for monitoring during intensive care. The electrodes used for ECG monitoring during intensive care must be suited for long term monitoring applications. The Tektronix silver/silver-chloride electrode system provided with the Type 410 Physiological Monitor is ideally suited

to this application as the electrode paste supplied produces no subject discomfort or skin irritation and the relatively large amount of paste required between the subject and the electrode prevents the paste from drying out due to evaporation and skin absorption.

blood
pressure

The second physiological parameter often of prime importance in intensive care monitoring is blood pressure. Blood pressure can be and often is monitored using an intra-arterial catheter and transducer; however, the catheter results in considerable subject discomfort and many intensive care units prefer to monitor blood pressure by some alternate method.

Korotkoff
system

Blood pressure can be monitored using the automatic cuff pump and Korotkoff microphone blood-pressure measurement system shown in Fig. 8-11. Although this system is occasionally used in intensive care units, it also possesses the disadvantage of being somewhat uncomfortable to the subject (bruises), and more importantly, being a sampling technique, it does not provide a continuous record of the subject's blood pressure. Thus, if for some reason the subject's blood pressure were to suddenly drop, this system may take a minute or so to detect this pressure drop.

plethysmo-
graph

Blood pressure monitoring with a plethysmograph offers the least discomfort to the subject; however, it provides only a relative indication of the well being of the circulatory system rather than providing absolute values for diastolic and systolic pressure.

Although considerable controversy still exists as to the best method for routine blood pressure measurement in an intensive care unit, each of these techniques possesses attributes and a single intensive care unit may employ one or more of these techniques and, indeed, all three may be available if required. Although diastolic and systolic arterial pressure are commonly monitored, mean arterial pressure and venous pressure are also monitored in some instances.

respiration	It is often desirable to monitor the subject's respiratory activity during intensive care; this may be accomplished with a thermistor pneumograph placed in the subject's nostril. It is often also
body temperature	desirable to monitor body temperature in intensive care subjects via a rectal or armpit thermistor probe.
pacemaker	Monitoring of the physiological signals referred to previously necessitates numerous electrodes, etc. being placed on the subject. In addition, it is often desirable to have cardiac pacemaker electrodes applied to the subject's chest. Although these electrodes are not used during routine intensive care, they should be connected to a cardiac pacemaker for immediate emergency use if required.

28.2 INTENSIVE CARE INSTRUMENTATION

A conceivable intensive care instrumentation system is shown in Figs. 28-1, 28-2 and 28-3. Fig. 28-1 depicts the total instrumentation layout in a four-bed intensive care unit. Each of the four beds includes separate subject-monitoring instrumentation providing an indication of the subject's physiological condition as indicated in Fig. 28-2. Signals from each of these four instrumentation modules are also connected to a central nurse's station to permit monitoring by the nurse on duty, to allow selective recording of the ECG, and to allow the ECG signal and/or audio-visual information to be transmitted throughout the hospital via a closed-circuit television link. The television camera and closed circuit link may be regarded as "luxury items," most other features shown are essential.

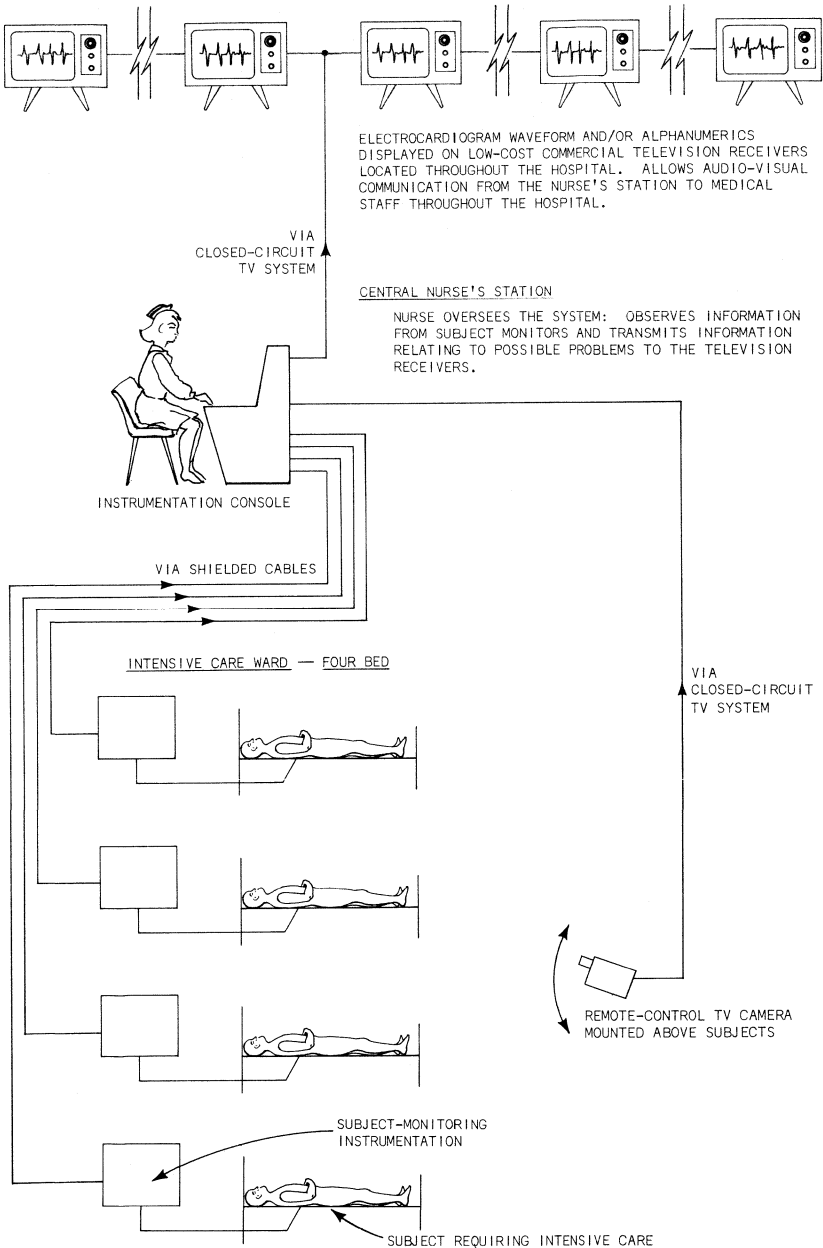


Fig. 28-1. Instrumentation in an intensive care unit.

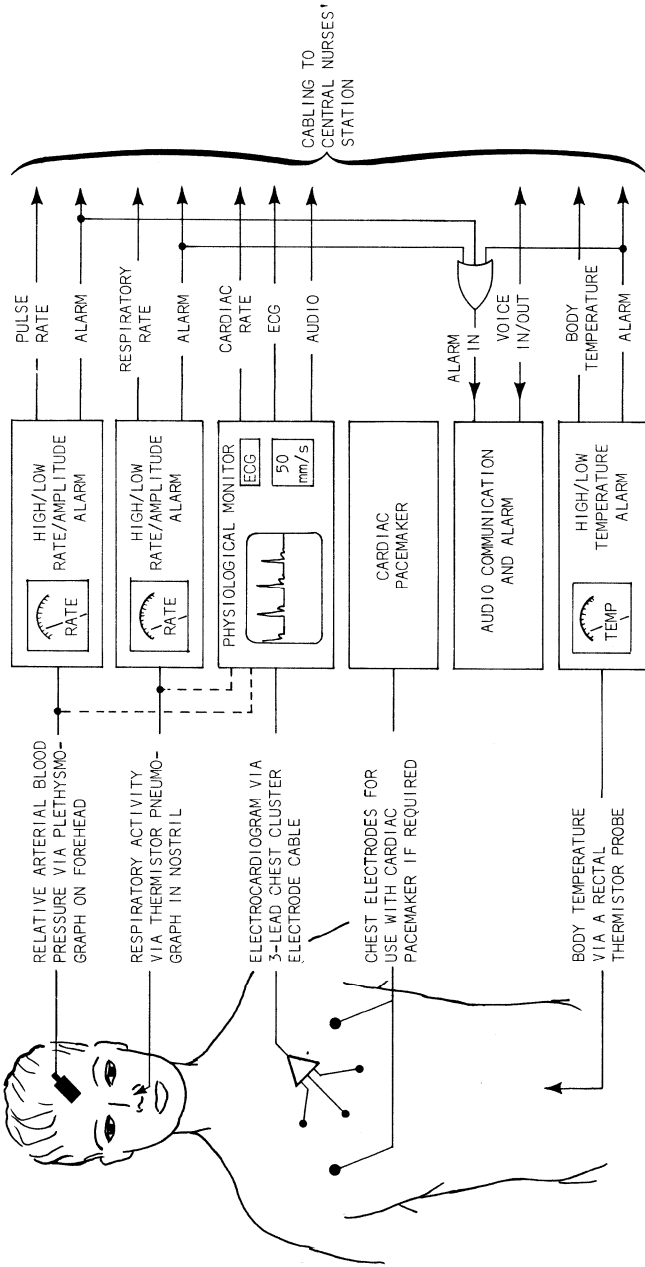


Fig. 28-2. Subject monitoring instrumentation for one intensive care "bed."

plethysmo-
graph

Referring to the subject-monitoring instrumentation located beside each intensive care bed as shown in Fig. 28-2, relative arterial blood pressure, respiratory activity, the electrocardiogram and body temperature are monitored. Relative arterial blood pressure is monitored via a plethysmograph on either the subject's forehead, his nasal septum or the lobe of his ear. Finger plethysmographs are rarely used in intensive care due to their susceptibility to subject movement. The instrumentation associated with the plethysmograph may provide an indication of cardiac rate; it may also provide an alarm signal should the cardiac rate go above or below preset limits or should the amplitude of signal produced by the plethysmograph fall below a predetermined level (indicating a loss in blood pressure or flow).

pneumo-
graph

Respiratory activity is monitored via a thermistor pneumograph in the nostril with the associated instrumentation providing an indication of respiratory rate as well as providing an alarm if this rate falls outside predetermined limits or if the signal level produced by the thermistor pneumograph is reduced below some predetermined amplitude (indicating a loss of respiratory activity).

Type 410
Monitor

The electrocardiogram is monitored by a Tektronix Type 410 Physiological Monitor using a Tektronix three-lead chest-cluster electrode cable. The physiological monitor provides an ECG output from a low impedance source as well as an audio output to audibly indicate the cardiac rate or to indicate a loss of cardiac activity.

thermistor
probe

Body temperature is monitored via a rectal thermistor probe. The associated instrumentation indicating body temperature may also contain an alarm system which will be activated if the body temperature should fall outside predetermined limits.

television
monitoring

Instrumentation located beside each intensive care bed should, preferably, be away from the subject's range of vision as its presence can be somewhat disconcerting to some subjects. The intensive care ward may also contain a closed-circuit television system to allow one or more subjects to be viewed via a television camera. This television camera may continuously scan the subjects in the intensive care ward or its position may be controlled from the central nurse's station.

additional
equipment

In addition to physiological and visual monitoring of the subject, a cardiac pacemaker module and audio communication and alarm module are also included with the instrumentation at each intensive care bed. The cardiac pacemaker provides variable-amplitude, variable-rate pulses for cardiac pacemaking should it be required. The audio communication and alarm panel provides audio-visual alarm indication of abnormalities in blood pressure, cardiac rate, respiratory activity or body temperature and provides audio communication between the intensive care bed and the nurse's station.

central
nurse's
station

An intensive care unit central nurse's station is shown in Fig. 28-3. Multiconductor cable connects the output from the four subject-monitoring sites located beside each intensive care bed to the central nurse's station. Each subject's ECG is continuously displayed via a four channel oscilloscope and is continuously recorded on a memory loop tape recorder. This tape recorder contains the previous one-minute ECG history for each subject by recording the ECG on a tape loop "one minute" in length. While some central stations duplicate all physiological indicators at the central station, this is normally unnecessary and, in the interest of simplicity, it is preferable to only provide one set of indicators for relative blood pressure, respiratory activity and body temperature. These indicators can be manually switched between the four beds or the switching may be activated by the alarm system with the monitors being automatically switched to the bed providing the alarm signal. When an alarm is received at the central nurse's station, it may also be used to connect the appropriate ECG signal to a scan converter and ECG chart recorder and to start the chart recorder. In this way a permanent record is achieved on the chart recorder beginning one minute prior to the alarm being sounded and information is displayed on the scan converter for transmission via the hospital closed-circuit TV system to other medical personnel involved. The scan converter and closed-circuit TV system may also incorporate alphanumeric input to allow alphanumeric data relating to the intensive care subject to be displayed on television receivers located throughout the hospital.

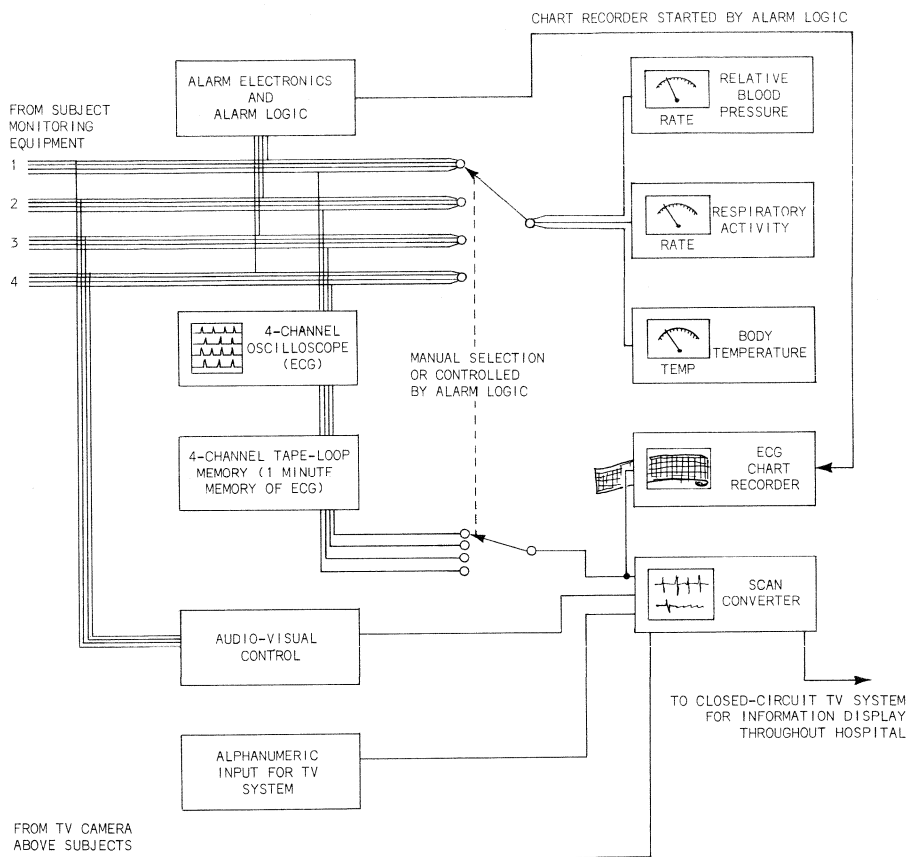


Fig. 28-3. Central nurses' station instrumentation for 4 intensive care beds.

computers
in
intensive
care

Increasing use is being made of digital computers in intensive care units. These computers provide storage of the subject's physiological data and can continuously interrogate this data to determine if it is within predetermined limits. The computer can perform multivariant analysis of the subject's ECG and can compare his ECG with previous records to indicate changes in the ECG which may be clinically significant. The advent of the time shared computer system allows one computer to simultaneously perform many tasks throughout a hospital, thus making the computer economically feasible for many hospitals.

SECTION IV

APPENDIX

Chapter 29 describes custom instrumentation required to perform the measurements discussed in Section II. While most of these custom items are simple and should more correctly be referred to as adapters, they are necessary for the correct operation of the measurement systems and are not generally available commercially.

Chapter 30 gives formal definitions for the biomedical terminology used throughout this book. It is hoped that this chapter contains the definitions of any terms used that would not normally be regarded as common terminology by the electronics engineer.

CUSTOM INSTRUMENTATION

This chapter provides circuits and construction details for various items of custom instrumentation that are required to implement some of the biophysical measurements discussed in Sections II and III. Details of these items are presented here since similar items do not appear to be available commercially. These items have been constructed and used satisfactorily for the measurement technique for which they were intended. Normal tolerance variations in components may, however, necessitate minor circuit changes in some instances. It is anticipated that the items presented will be constructed by competent electronic technicians or electronic engineers who are able to evaluate the performance of the device and modify this performance if necessary.

Although the Tektronix, Inc. part number is provided for all the parts necessary to construct these items, most of the parts are common electronic components and are available through local suppliers. We wish to encourage purchase from these local suppliers whenever possible. If necessary, however, parts may be purchased in small quantities through Tektronix field offices located throughout the United States or through Tektronix subsidiaries or distributors in other countries. These field offices, subsidiaries and distributors are listed in the current Tektronix catalog. When purchasing these parts from Tektronix, please quote the Tektronix part numbers for the individual components shown with the circuits. While all parts were available from Tektronix at the time of preparation of this text, Tektronix makes no guarantee as to the continued availability of these parts.

The items described in Sections 29.1 through 29.5 are intended for use in conjunction with the Tektronix Type 410 Physiological Monitor.

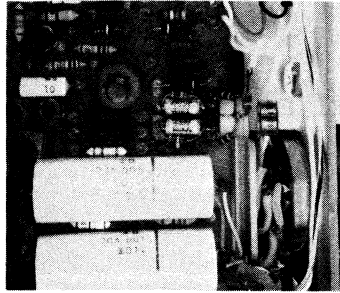
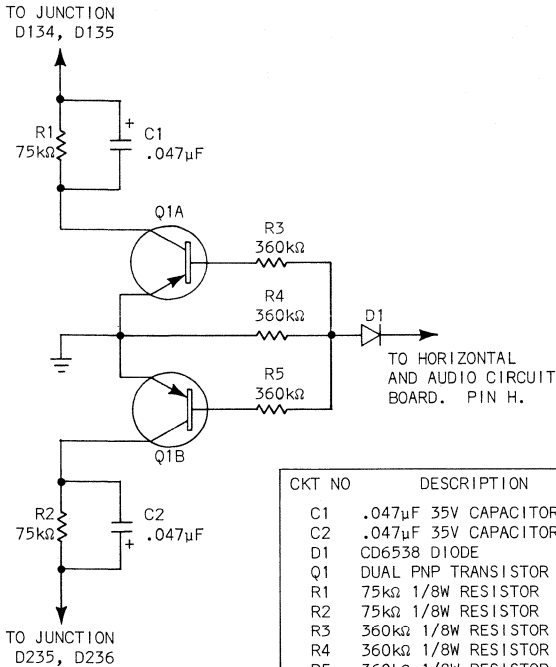
The items described in Sections 29.6 through 29.9 are intended for use with the Tektronix Type 3A8 Operational Amplifier plug-in unit. These items may also be used in conjunction with the Tektronix Type 0 Operational Amplifier plug-in unit. It is also probable that these circuits may be suited for use with other operational amplifier modules.

The items described in Sections 29.10 through 29.15 are independent items of instrumentation and are not intended for use with any specific instrument.

29.1 TYPE 410 MODIFICATION FOR FETAL ECG USE

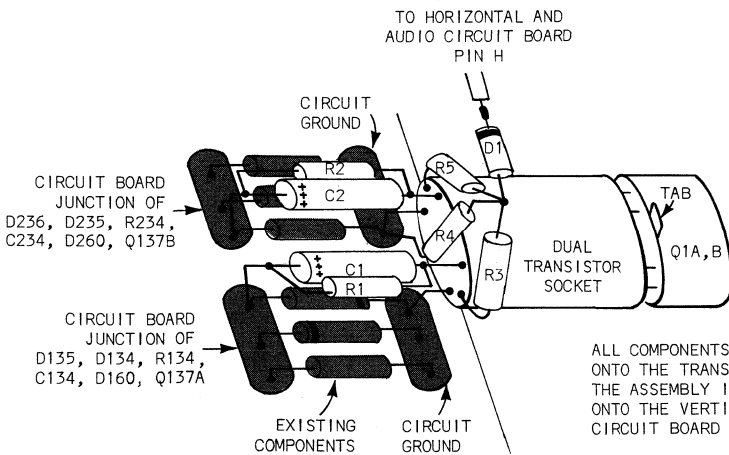
Monitoring of the fetal ECG requires the use of a physiological monitor with an amplifier having a low frequency 3 dB point of >1 Hz. The standard 410 Physiological Monitor has a low frequency response of <0.1 Hz and is thus unsuited for monitoring the fetal ECG. The EEG function of the Type 410 Physiological Monitor may be modified to allow it to be used to record the fetal ECG. This modification consists of modifying the low frequency response of the Vertical Amplifier in the EEG position from <0.1 Hz to >1 Hz. See Fig. 29-1.

The low frequency response of the amplifier is changed by decreasing the time constant of the AC coupling network from 1 microfarad/2.2 megohm to 1 microfarad/75 kilohm by adding 75 kilohm resistors when the 410 is operated in the EEG mode. This also reduces the amplifier's gain by a factor of 2. Since no spare switch positions are available on the 410 INPUT SELECTOR switch, switching of the 75 kilohm resistors must be accomplished using transistors. The 410 horizontal and audio circuit produces a positive logic level when the 410 is operated in the EEG mode. This level is used to freerun the sweep and to disable the audio circuit, however, it may also be used in this modification to turn on the switching transistors Q1A and Q1B.



ADDITIONAL COMPONENTS LOCATED WITHIN THE 410.

CKT NO	DESCRIPTION	TEKTRONIX PN
C1	.047μF 35V CAPACITOR	290-0282-00
C2	.047μF 35V CAPACITOR	290-0282-00
D1	CD6538 DIODE	152-0185-00
Q1	DUAL PNP TRANSISTOR	151-0261-00
R1	75kΩ 1/8W RESISTOR	317-0753-00
R2	75kΩ 1/8W RESISTOR	317-0753-00
R3	360kΩ 1/8W RESISTOR	317-0364-00
R4	360kΩ 1/8W RESISTOR	317-0364-00
R5	360kΩ 1/8W RESISTOR	317-0364-00
--	DUAL TRANSISTOR SOCKET	136-0235-00
--	8 INCH INSULATED WIRE	--



ALL COMPONENTS ARE MOUNTED ONTO THE TRANSISTOR SOCKET. THE ASSEMBLY IS THEN WIRED ONTO THE VERTICAL AMPLIFIER CIRCUIT BOARD WITHIN THE 410.

Fig. 29-1. Type 410 modification for fetal ECG use.
EEG position now 5 mm/50 μV, 1 Hz — 100 Hz.

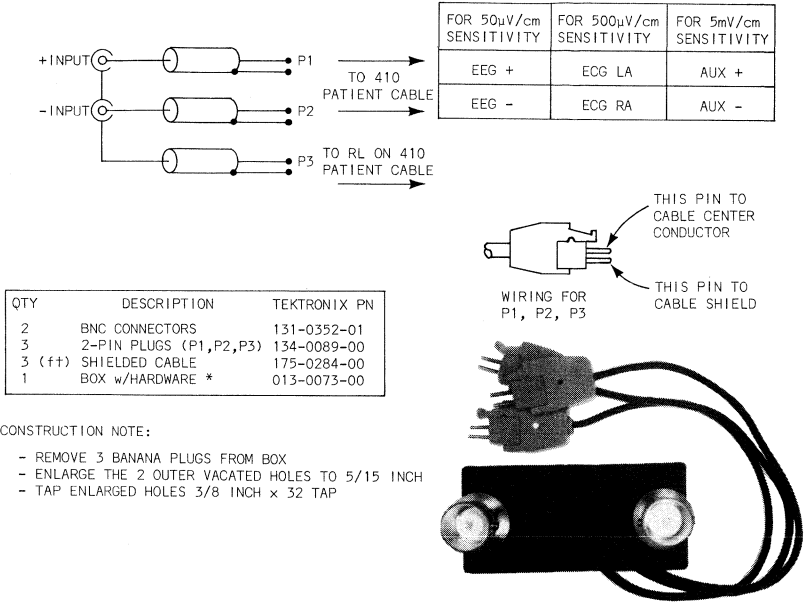


Fig. 29-2. A BNC input adapter for the Type 410 Monitor.

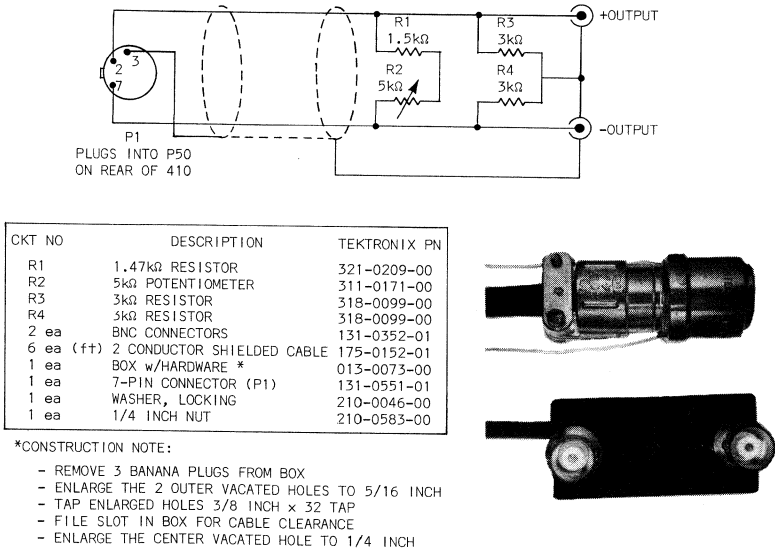


Fig. 29-3. A signal output adapter for the Type 410 Monitor that provides a differential signal output from the Type 410 Monitor at 100 mV for each cm deflection on the 410 CRT.

29.2 A BNC INPUT ADAPTER FOR THE TYPE 410 MONITOR

Refer to Fig. 29-2. This adapter requires little explanation, being entirely passive. The adapter simply provides BNC to "410 System" adaption to allow conventional BNC accessories, etc., to be used to connect physiological signals to the Type 410 Monitor.

Referring to the markings located on the Type 410 patient cable, the adapter may be connected to the AUX+ and AUX- positions to allow the 410 to be used in the auxiliary mode, providing a sensitivity of 5 mV per centimeter. With the adapter connected to the ECG LA and ECG RA positions, the 410 must be operated in the ECG lead I mode and will provide a sensitivity of 500 μ V per centimeter. With the adapter connected to the EEG+ and EEG- positions, the Type 410 must be operated in the EEG mode and will provide a sensitivity of 50 μ V per centimeter.

29.3 A SIGNAL OUTPUT ADAPTER FOR THE TYPE 410 MONITOR

Refer to Fig. 29-3. This adapter provides a vertical signal output from the Type 410 Vertical Amplifier via conventional BNC connectors. A differential output of 100 mV for each centimeter of vertical deflection on the Type 410 CRT is provided. The adapter should, preferably, be used with auxiliary equipment having a differential input, however satisfactory results may be obtained when using only one output from the adapter in conjunction with equipment having only a "single ended" input. The sensitivity in the single ended mode will be 50 mV for each centimeter of vertical deflection and care should be taken to avoid excessive common mode signals at the 410 input.

The Type 410 provides a differential vertical signal out at the rear of the instrument of approximately 65 μ A for each centimeter of CRT deflection. A differential output voltage is produced across a load on this current source consisting of R1, R2, R3 and R4. R2 varies this load to allow the output voltage to be varied. If alternative output voltages are required the values of R3 and R4 should be changed accordingly.

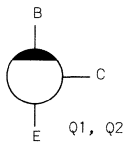
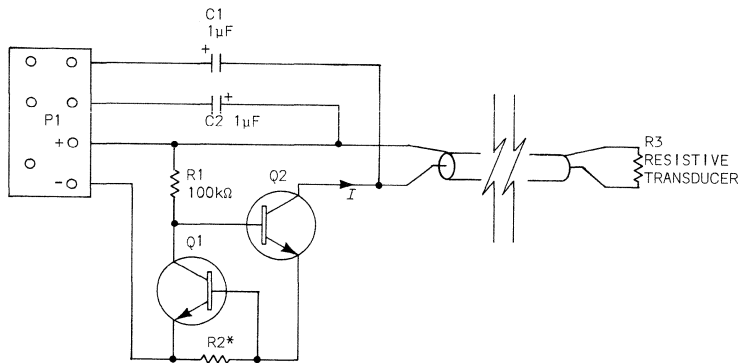
29.4 A RESISTIVE TRANSDUCER ADAPTER FOR THE TYPE 410 MONITOR

Refer to Fig. 29-4. This adapter allows resistive transducers to be used with the Type 410 Monitor operating in the auxiliary mode. A change in the resistive transducer's nominal resistance of less than 0.1% produces 1 centimeter of vertical deflection on the Type 410 CRT. The frequency response of the system is from 0.1 Hz to 250 Hz.

R1, R2, Q1 and Q2 provide a constant current source for the resistive transducer, thus the voltage across the transducer is proportional to the transducer's resistance. Capacitors C1 and C2 couple changes in this voltage to the vertical amplifier of the Type 410 Monitor.

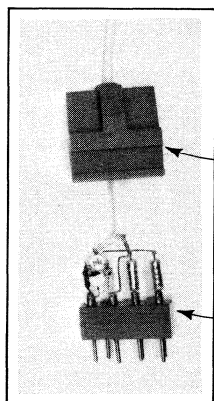
Almost any resistive transducer having a nominal resistance below 10 k Ω may be used, however the adapter is not intended for use with bridge type resistive transducers. The adapter was specifically constructed for use with the thermistor pneumograph described in Section 29.5.

TRANSDUCER SUPPLY CURRENT	VALUE OF R2	TEKTRONIX PN FOR R2	OPTIMUM TRANSDUCER NOMINAL RESISTANCE	SENSITIVITY CHANGE IN RESISTANCE (δR) FOR 1 DIV VERTICAL DEFLECTION ON 410 CRT
$I \approx 1\text{mA}$	680 Ω	317-0681-00	10k Ω - 5k Ω	$\delta R \approx 5\text{ OHMS/DIV}$
2mA	330 Ω	317-0331-00	5k Ω - 2k Ω	2.5 OHMS/DIV
5mA	130 Ω	321-0108-00	2k Ω - 1k Ω	1 OHM/DIV
10mA	62 Ω	317-0620-00	LESS THAN 1k Ω	0.5 OHM/DIV



CKT NO	DESCRIPTION	TEKTRONIX PN
C1	1 μ F 35V CAPACITOR	290-0308-00
C2	1 μ F 35V CAPACITOR	290-0308-00
P1	7-PIN PLUG	134-0090-00
Q1	SILICON NPN	151-0206-00
Q2	SILICON NPN	151-0206-00
R1	100k Ω 1/8W RESISTOR	317-0104-00
R2	* 1/8W RESISTOR	*
R3	RESISTIVE TRANSDUCER *	--
QTY	4 ft SHIELDED CABLE	175-0284-00

*SEE CHART FOR VALUE AND PN



COVER, PART OF P1, SNAPS OVER PLUG AND COMPONENTS

P1

ALL COMPONENTS ARE MOUNTED ONTO THE PINS OF THE 7-PIN PLUG. THE PLUG COVER IS THEN FITTED LEAVING THE SHIELDED CABLE PROJECTING FOR CONNECTION TO A RESISTIVE TRANSDUCER.

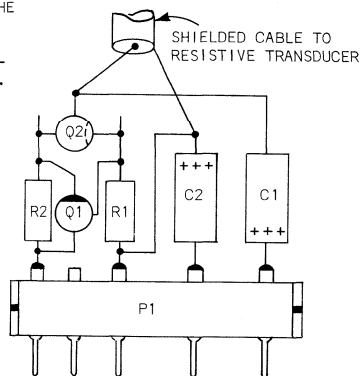
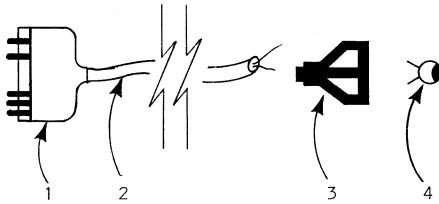


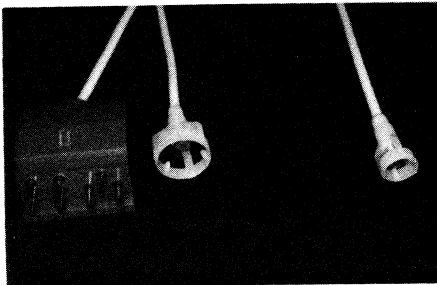
Fig. 29-4. A resistive transducer adapter for the Type 410 Monitor.



- SOLDER THERMISTOR TO CABLE
- SLIDE HOLDER OVER JUNCTION
- FIX HOLDER IN PLACE WITH CEMENT
- SOLDER CABLE TO TRANSDUCER ADAPTER

ITEM	DESCRIPTION	QTY	TEKTRONIX PN
1	RESISTIVE TRANSDUCER INTERFACE ADAPTER - 5mA.	1	REFER FIG 29-4 FOR DETAILS
2	PVC-COVERED COAXIAL CABLE	4 (ft)	175-0284-00
3	HOLDER - LARGE 7/16 INCH DIA OR SMALL 1/4 INCH DIA	1	352-0077-00 352-0078-00
4	1kΩ THERMISTOR	1	307-0127-01

THERMISTOR PNEUMOGRAPH
w/LARGE HOLDER
(FOR ADULTS)



OPTIONAL SMALL
HOLDER
(FOR CHILDREN)

THERMISTOR PNEUMOGRAPH
FITTED TO SUBJECT

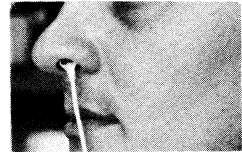


Fig. 29-5. A thermistor pneumograph for the Type 410 Monitor.

29.5 A THERMISTOR PNEUMOGRAPH FOR THE TYPE 410 MONITOR

Refer to Fig. 29-5. This pneumograph allows a subject's respiration to be monitored by monitoring the temperature difference between inhaled and exhaled air with a thermistor.

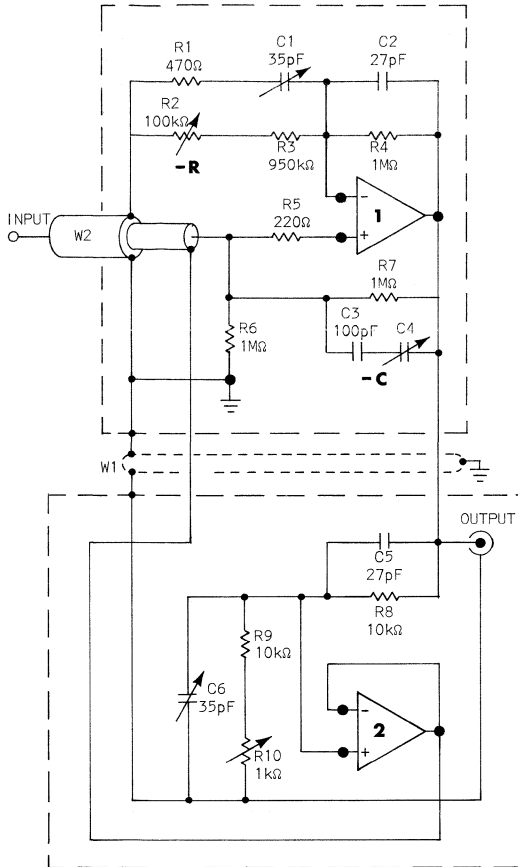
A thermistor having a nominal resistance of $1\text{ k}\Omega$ is used in conjunction with suitable mechanical holders and the resistive transducer adapter described in Section 29.4. Either a large holder (for adults) or a small holder (for children) may be used to hold the pneumograph in place at the subject's nostril.

29.6 AN INPUT NEUTRALIZING ADAPTER FOR THE TYPE 3A8 OPERATIONAL AMPLIFIER

Refer to Fig. 29-6. This adapter provides the extremely high input impedance necessary to allow amplification of signals derived from microelectrodes. The input resistance into this adapter is greater than 10^9 ohms, the input capacitance is less than 1 picofarad and the adapter has a gain of 2X and a low output impedance to allow it to be used in conjunction with conventional instrumentation. The adapter uses both operational amplifier units available in the Tektronix Type 3A8 Operational Amplifier plug-in unit.

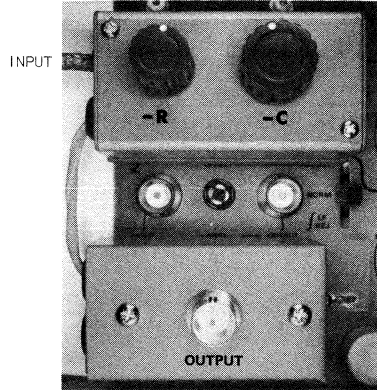
Operational amplifier #1 is operating as a non-inverting amplifier having a gain of 2 determined by R2, R3 and R4. R1, C1 and C2 provide frequency response compensation for these gain setting resistors. Positive feedback is produced via R7, C3 and C4 to a current summing point at the junction of R5 and R6. This positive feedback current is adjusted so as to equal the sum of the input current to the operational amplifier via R5 plus the current through R6. Thus the current from the input must be zero. The output from operational amplifier #1 provides the output from the adapter and also provides an input to operational amplifier #2 operating as a unity gain follower. The components associated with the positive input of operational amplifier #2 form a 2X attenuator, thus the output from operational amplifier #2 is half of the output of operational amplifier #1, that is, equal to the input to operational amplifier #1. The output from operational amplifier #2 is used to drive an ungrounded shield surrounding the input to the adapter. This provides input "guarding" to effectively reduce the capacitance of the input cable.

REFER ALSO TO 3A8 INSTRUCTION MANUAL.



$R_i > 10^9\Omega$
 $C_i < 1pF$
 GAIN 2X

SET 3A8 OPERATIONAL AMPLIFIERS Z_i AND Z_f SWITCHES TO EXT, GRID SELECTION TO + GRID.



SET C1 FOR OPTIMUM TRANSIENT RESPONSE WITH INPUT CONNECTED DIRECTLY TO 40mV CALIBRATOR.

SET C6 AND R10 FOR OPTIMUM TRANSIENT RESPONSE WITH INPUT CONNECTED TO 40mV CALIBRATOR VIA 10MΩ SERIES RESISTANCE.

R2 AND C4 ARE FRONT PANEL ADJUSTMENTS.

CKT NO	DESCRIPTION	TEKTRONIX PN	MECHANICAL QTY	DESCRIPTION	TEKTRONIX PN
C1	9-35pF CAPACITOR	281-0092-00			
C2	27pF CAPACITOR	283-0076-00		1 ADAPTER-TERMINAL ASSY	013-0048-01
C3	100pF CAPACITOR	283-0060-00		1 COMPENSATION ADAPTER *	013-0081-00
C4	VAR CAPACITOR (-C)	-- *		1 1/4 INCH GROMMET	348-0020-00
C5	27pF CAPACITOR	283-0076-00		3 5/16 INCH GROMMETS	348-0003-00
C6	9-35pF CAPACITOR	281-0092-00		1 CONNECTOR BNC	131-0106-01
R1	470Ω 1/4W 5% RESISTOR	315-0471-00		1 KNOB FOR R2	366-0153-00
R2	100kΩ POTENTIOMETER (-R)	311-0467-00		1 NUT FOR R2	210-0583-00
R3	950kΩ 1/8W 1% RESISTOR	318-0095-00		1 WASHER-LOCKING FOR R2	210-0046-00
R4	1MΩ 1/8W 1% RESISTOR	321-0481-00		1 LUG-SOLDER FOR R2	210-0223-00
R5	220Ω 1/4W 5% RESISTOR	315-0221-00			
R6	1MΩ 1/8W 1% RESISTOR	321-0481-00			
R7	1MΩ 1/8W 1% RESISTOR	321-0481-00			
R8	10kΩ 1/8W 1% RESISTOR	321-0289-00			
R9	10kΩ 1/8W 1% RESISTOR	321-0289-00			
R10	1kΩ POTENTIOMETER	311-0635-00			
W1	1 ft 2-CONDUCTOR SHIELDED CABLE	175-0072-00			
W2	1 ft 1-CONDUCTOR SHIELDED CABLE	175-0284-00			
	1 ft SHIELDING BRAID WIRE	176-0047-00			

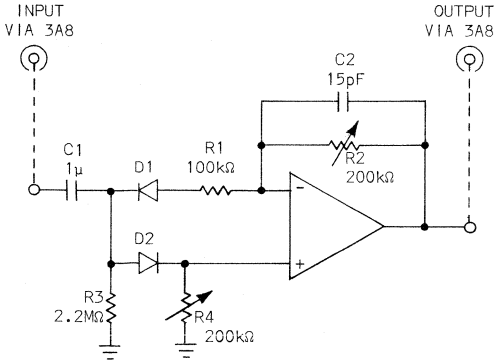
*C4 AND ASSOCIATED HARDWARE ARE SUPPLIED AS PART OF THE COMPENSATING ADAPTER. THE TEKTRONIX PN FOR C4 IS 281-0090-00.

Fig. 29-6. An input neutralizing adapter for the Type 3A8 Operational Amplifier.

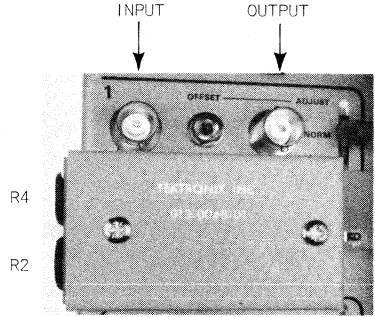
29.7 AN ABSOLUTE VALUE ADAPTER FOR THE TYPE 3A8 OPERATIONAL AMPLIFIER

Refer to Fig. 29-7. This adapter provides full wave rectification of input signals producing positive output signals for both positive and negative input signals. The adapter has an absolute gain of 1.

With a negative going input, diode D1 conducts and the operational amplifier operates as an inverting amplifier with unity gain determined by R1, R2, R3 and the output impedance of the input source. With a positive going input, diode D2 conducts and the operational amplifier operates as a unity gain, noninverting amplifier, as " Z_i " at the operational amplifier negative input is effectively infinite. R4 ensures that the adapter has the same input resistance for both positive and negative signals, thus maintaining a constant potential on C1. Diodes D1 and D2 effectively reduce the input voltage level by approximately 0.3 volts.

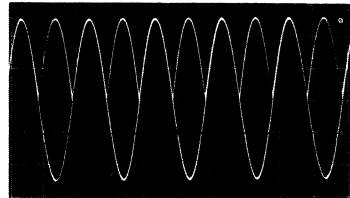


ADJUST R4 FOR 50% DUTY FACTOR WITH SINEWAVE
 ADJUST R2 FOR SYMMETRY WITH SINEWAVE

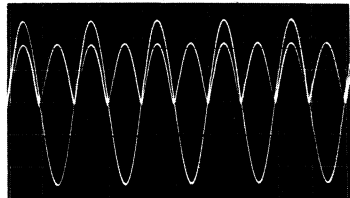


CKT NO	DESCRIPTION	TEKTRONIX PN
C1	1µF 100V CAPACITOR	285-0815-00
C2	15pF CAPACITOR	281-0509-00
D1 } D2 }	MATCHED PAIR DIODES	152-0110-00
R1	100kΩ 1/4W RESISTOR	315-0104-00
R2	200kΩ POTENTIOMETER	311-0660-00
R3	2.2MΩ 1/4W RESISTOR	316-0225-00
R4	200kΩ POTENTIOMETER	311-0660-00
MECHANICAL		
QTY		
1	ADAPTER TERMINAL ASSY	013-0048-01
2	5/16 INCH GROMMETS	348-0003-00

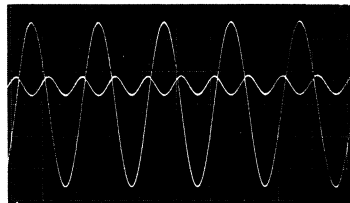
PERFORMANCE AT 500HZ



10V P-P INPUT OUTPUT = |INPUT|



1V P-P INPUT OUTPUT = 0.75 |INPUT|

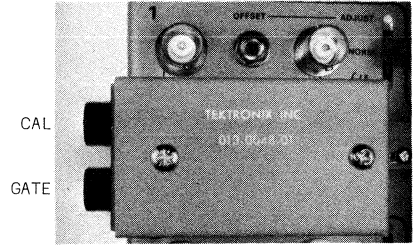
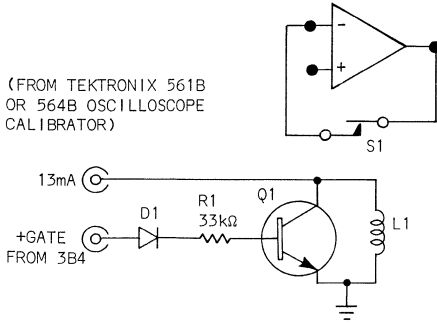


0.1V P-P INPUT
 OUTPUT = 0.25 |INPUT| + 5mV

Fig. 29-7. An absolute value adapter for the Type 3A8 Operational Amplifier.

REED RELAY (S1) OPENS AND CLOSES
IN LESS THAN 1ms
POSITIVE GATING PULSE OPENS S1

(FROM TEKTRONIX 561B
OR 564B OSCILLOSCOPE
CALIBRATOR)



CKT NO	DESCRIPTION	TEKTRONIX PN
D1	DIODE - SILICON	152-0185-00
L1	REED DRIVE COIL	108-0442-00
Q1	2N1893 TRANSISTOR	151-0096-00
R1	33kΩ 1/4W RESISTOR	315-0333-00
S1	REED SWITCH	260-0877-00
MECHANICAL		
QTY		
1	ADAPTER TERMINAL ASSY 013-0048-01	

PERFORMANCE AS GATED INTEGRATOR - .5ms/div
INTEGRATOR - $Z_i = 0.1M\Omega$, $Z_f = 0.1\mu F$
INTEGRATOR GATED "ON" BY 3B4 FOR 2.3ms
INPUT TO INTEGRATOR 1V DC

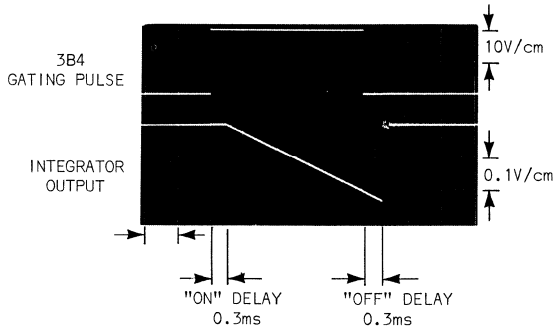
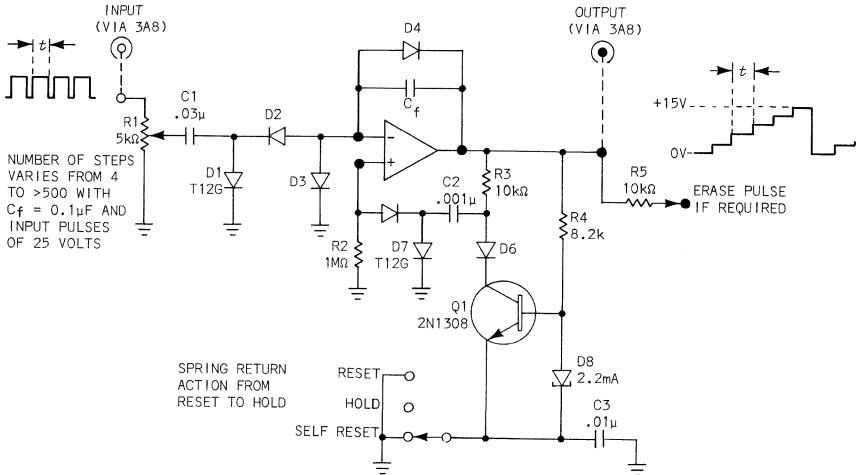


Fig. 29-8. A low-speed gating adapter for the Type 3A8 Operational Amplifier.

29.8 A LOW SPEED GATING ADAPTER FOR THE TYPE 3A8 OPERATIONAL AMPLIFIER

Refer to Fig. 29-8. This adapter provides a reed switch between the negative input and output of an operational amplifier for various gating applications, particularly for gated integration applications. The reed drive coil, L1, is powered from the 13 mA available when the calibrator in a Tektronix Type 561B or 564B Oscilloscope is operated in the 40 volt DC position. With no input signal, this 13 mA flows in L1, causing S1 to close. When a positive level is applied to the base of Q1, Q1 conducts, diverting the 13 mA to Q1 and thus causing S1 to open. R1 is chosen as 33 k Ω to allow the adapter to be powered from the 20 volt plus gate signal available from the Type 3B4 Time Base unit. This adapter may be used in conjunction with lower voltage gating signals by correspondingly reducing the value of R1. A delay of 0.3 ms is produced between the application of the gating pulse and the activation of S1.



CKT NO	DESCRIPTION	TEKTRONIX PN
C1	.03μF 100V CAPACITOR	285-0702-00
C2	.001μF 200V CAPACITOR	283-0067-00
C3	.01μF 200V CAPACITOR	283-0005-00
Cf	SELECTED BY TYPE 3A8 Zf CONTROL (0.1μF)	
D1	DIODE T12G	152-0008-00
D2-D6	5 EACH S1 DIODE CD61165	5 EACH 152-0245-00
D7	DIODE T12G	152-0008-00
D8	2.2mA TUNNEL DIODE TD-2	152-0402-00
R1	5kΩ MIN POT	311-0310-00
R2	1MΩ 1/4W RESISTOR	315-0105-00
R3	10kΩ 1/4W RESISTOR	315-0103-00
R4	8.2kΩ 1/4W RESISTOR	315-0822-00
R5	10kΩ 1/4W RESISTOR	315-0103-00
Q1	2N1308 TRANSISTOR	151-0072-00
MECHANICAL		
QTY		
1	ADAPTER TERMINAL ASSY	013-0048-01
1	3-POSITION SWITCH	260-0614-00
1	KNOB FOR R1	366-0189-00
2	WASHERS FOR R1 AND SWITCH	210-0905-00
2	NUTS FOR R1 AND SWITCH	210-0562-00

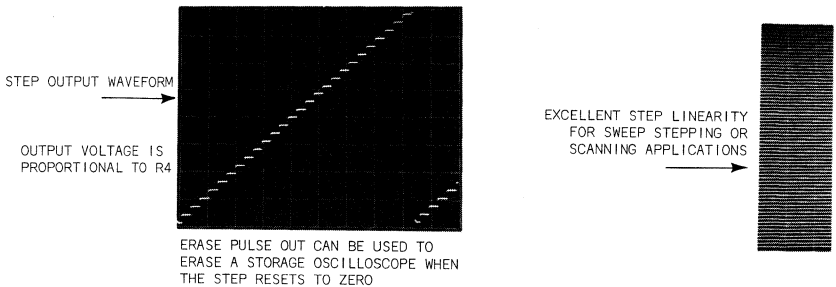
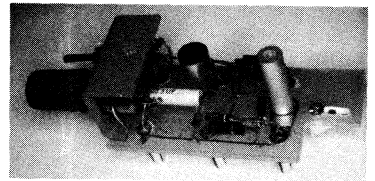
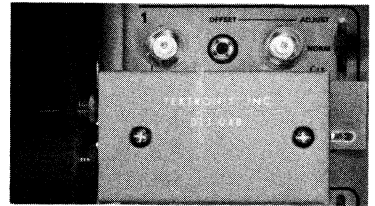


Fig. 29-9. A resetting step generator adapter for the Type 3A8 Operational Amplifier.

29.9 A RESETTING STEP GENERATOR ADAPTER FOR THE TYPE 3A8 OPERATIONAL AMPLIFIER

Refer to Fig. 29-9. This adapter produces a staircase waveform, the stepping commands being received from an external input. This external input may typically be a gating pulse derived from the time base of an oscilloscope. The output staircase may also be used to erase a storage oscilloscope during step reset.

The circuit shown provides a step ramp of 15 volts peak, with the number of steps continuously variable from 4 to 500 or more. The required step command input is a 20 to 25 volt gate waveform from a source impedance of 5,000 ohms or less. Component selection allows for the use of nearly any step command waveform greater than about 1.5 volts. Linearity appears to be better than $\pm 3\%$.

A negative charge is applied to D2 from the input, the charge is then transferred to C_f by the operational amplifier to provide a voltage step at the output. The charge is the charge in C1 developed during the "on" time of the input pulse. The charge is transferred to C_f during the off time of the input pulse. The amount of voltage developed across C_f for each step is dependent on the input pulse amplitude, the setting of R1 and the ratio $C1/C_f$.

The positive going staircase at the output provides both collector voltage for Q1 and current for the tunnel diode D8 which is normally in the low voltage state. When the staircase reaches the amplitude at which the current through R4 and D8 reaches 2.2 mA, the tunnel diode switches to its high voltage state, turning on Q1. The collector of Q1 snaps sharply negative. This negative going step is coupled to the +grid of the operational amplifier via C2 and D5, causing the output to fall negative. The negative transition at the output is also coupled via C2 and D5 to the positive grid, completing a regenerative loop and forcing the output to continue falling even when D8 has reverted to its low voltage state and Q1 is turned off. D8 and Q1, then, only trigger the reset; the "work" is done by the operational amplifier.

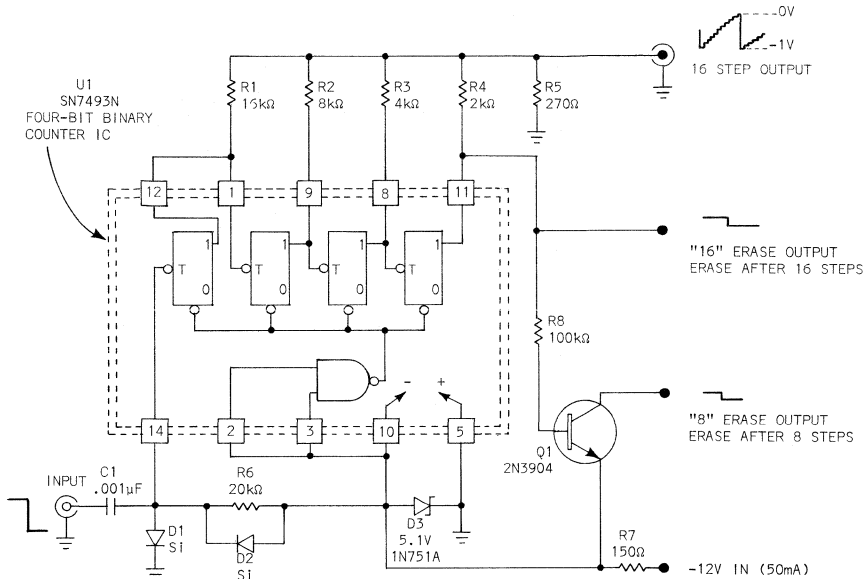
D3 prevents the falling output from driving the -grid negative and cancelling the regenerative action; D4 catches the output when it reaches ground potential. D6 prevents the collector of Q1 from being driven negative.

The negative transition drives the junction of C2 and R3 to approximately -15 volts, it then recovers to ground potential via the C2-R2 time constant. Until the +grid recovers to 0 volts and D5 disconnects, further input pulses to the -grid are shunted to ground via D3 and do not charge the stairstep. This provides a natural holdoff action. The circuit comes out of holdoff with a sharp regenerated step to 0 V.

29.10 A SELF-CONTAINED RESETTING STAIRSTEP GENERATOR

Refer to Fig. 29-10. This stairstep generator provides a 16 step ramp of 1 volt peak, the step command input requiring a negative going transition of greater than 2 volts in amplitude to initiate the stepping action. This input may typically be the gating pulse derived from the time base of an oscilloscope. The output stairstep may also be used to erase a storage oscilloscope during step reset or to erase one half of the screen of a split screen oscilloscope after 8 steps have been completed.

U1 consists of 4 toggling flip-flops with the "1" output of each flip-flop driving the clock input of the adjacent flip-flop. R1, R2, R3, R4 and R5 form a digital to analog converter to allow the digital levels produced by the 4 flip-flops to be transformed to analog steps. R5 may be reduced in value to provide less ramp output. A decrease in R5 will also improve the linearity of the circuit. The linearity of the circuit may also be improved by selecting U1 for optimum linearity.



STEP GENERATOR WILL STEP ONCE FOR EACH NEGATIVE TRANSITION. MINIMUM AMPLITUDE 2V.

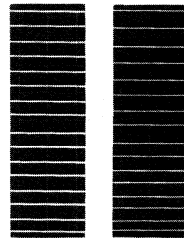
CKT NO	DESCRIPTION	TEKTRONIX PN
C1	.001μF 500V CAPACITOR	283-0078-00
D1	SILICON DIODE	152-0185-00
D2	SILICON DIODE	152-0185-00
D3	1N751A ZENER DIODE	152-0279-00
Q1	2N3904 TRANSISTOR	151-0190-00
R1	16.2kΩ ±1% RESISTOR	321-0309-00
R2	7.96kΩ ±1% RESISTOR	321-0638-00
R3	4.02kΩ ±1% RESISTOR	321-0251-00
R4	2.0kΩ ±1% RESISTOR	321-0222-00
R5	270Ω ±5% RESISTOR	315-0271-00
R6	20kΩ ±5% RESISTOR	315-0203-00
R7	150Ω 1/2W RESISTOR	301-0151-00
R8	100kΩ ±5% RESISTOR	315-0104-00
U1	INTEGRATED CKT T1 SN7493N	156-0032-00

MECHANICAL QTY	DESCRIPTION	TEKTRONIX PN
1	BOX - TERMINAL	202-0054-00
1	CONNECTOR - BNC MALE	131-0428-00
1	CONNECTOR - BNC FEMALE	131-0106-00
3	CONNECTOR - TERMINAL JACK	131-0251-00
1	SOCKET - INTEGRATED CKT	136-0269-00
1	COVER - BOTTOM	200-0252-00
2	SETScrew 4-40	213-0048-00
2	SCREW, THREAD-FORMING	213-0141-00

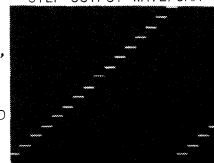
WITH R5 AT 270Ω THE OUTPUT IS ≈ 1V BUT IS SLIGHTLY NONLINEAR.

WITH R5 AT 27Ω (TEKTRONIX PN 315-0270-00) THE OUTPUT IS REDUCED TO = 0.1V, HOWEVER, LINEARITY IS IMPROVED. R8, Q1 AND "8" ERASE OUTPUT ONLY REQUIRED WITH SPLIT-SCREEN STORAGE OSCILLOSCOPES SUCH AS TYPE 564B.

STEP LINEARITY WITH 0.1V OUT R5 = 27Ω	STEP LINEARITY WITH 1V OUT R5 = 270Ω
---------------------------------------	--------------------------------------



STEP OUTPUT WAVEFORM



IF ONLY 8 STEPS REQUIRED, DISCONNECT [1] FROM [2] AND CONNECT IT TO [4].

LINEARITY MAY BE IMPROVED BY SELECTING U1.

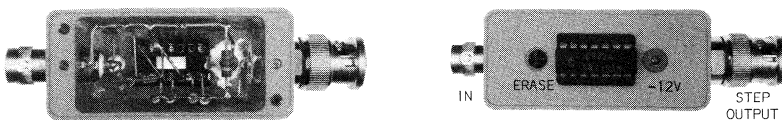
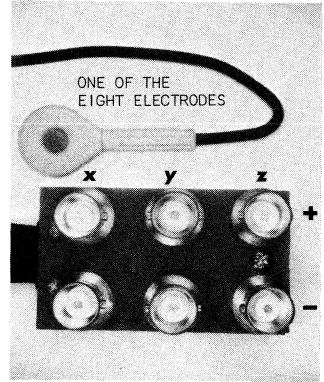
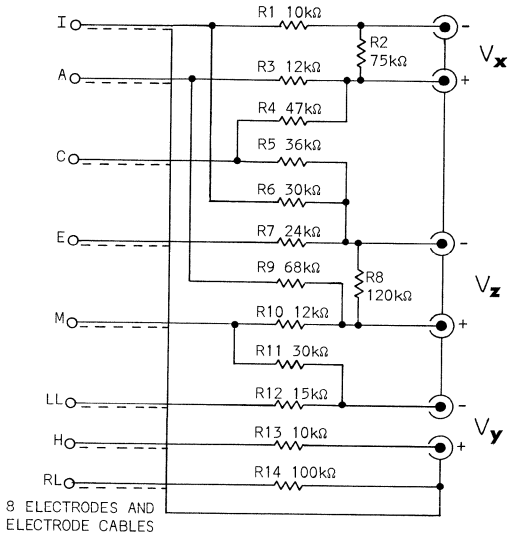


Fig. 29-10. A self-contained resetting stair generator.



CKT NO	DESCRIPTION	TEKTRONIX PN
R1	10kΩ 1/8W RESISTOR	317-0103-00
R2	75kΩ 1/8W RESISTOR	317-0753-00
R3	12kΩ 1/8W RESISTOR	317-0123-00
R4	47kΩ 1/8W RESISTOR	317-0473-00
R5	36kΩ 1/8W RESISTOR	317-0363-00
R6	30kΩ 1/8W RESISTOR	317-0303-00
R7	24kΩ 1/8W RESISTOR	317-0243-00
R8	120kΩ 1/8W RESISTOR	317-0124-00
R9	68kΩ 1/8W RESISTOR	317-0683-00
R10	12kΩ 1/8W RESISTOR	317-0123-00
R11	30kΩ 1/8W RESISTOR	317-0303-00
R12	15kΩ 1/8W RESISTOR	317-0153-00
R13	10kΩ 1/8W RESISTOR	317-0103-00
R14	100kΩ 1/8W RESISTOR	317-0104-00
MECHANICAL		
QTY		
1	ELECTRODE WITH 4ft CABLE	012-0121-21
A	ELECTRODE WITH 4ft CABLE	012-0121-21
C	ELECTRODE WITH 4ft CABLE	012-0121-21
E	ELECTRODE WITH 4ft CABLE	012-0121-21
M	ELECTRODE WITH 4ft CABLE	012-0121-21
LL	ELECTRODE WITH 6ft CABLE	012-0121-22
H	ELECTRODE WITH 4ft CABLE	012-0121-21
RL	ELECTRODE WITH 6ft CABLE	012-0121-25
6 ea	BNC CONNECTORS	131-0352-01
1 ea	ADAPTER-TERMINAL ASSY	013-0048-01*
1 ea	1/2 INCH GROMMET	348-0005-00

***CONSTRUCTION NOTE:**

- REMOVE 6 BANANA PLUGS FROM TERMINAL ASSY
- ENLARGE VACATED HOLES TO 3/8 INCH
- EPOXY 6 BNC CONNECTORS INTO HOLES
- REMOVE 2 BANANA SOCKETS FROM TERMINAL ASSY
- ENLARGE AND JOIN VACATED HOLES FOR 1/2 INCH GROMMET
- REMOVE PLUGS FROM THE ENDS OF THE 8 ELECTRODES

Fig. 29-11. A "Frank" network for vectorcardiographic use.

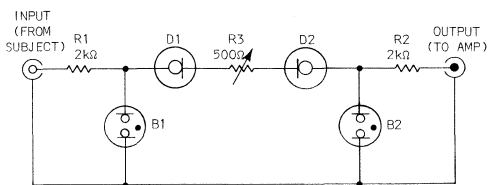
29.11 A FRANK NETWORK FOR VECTORCARDIOGRAPHIC USE

Refer to Fig. 29-11. This network provides the necessary electrode signal interconnections and attenuations for vectorcardiographic use using the Frank system. Eight silver/silver-chloride electrodes are applied to the subject and the three orthogonal vectorcardiographic signals, x , y , and z , are available as differential signals via three pairs of BNC connectors.

29.12 A CURRENT LIMITING ADAPTER FOR PROTECTION FROM ELECTRIC SHOCK

Refer to Fig. 29-12. This adapter limits current between the input and output or between the output and input to less than 300 μA peak, thus allowing conventional electronic instrumentation to be used in conjunction with human subjects.

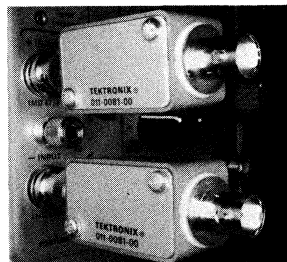
Current limiting is achieved by field effect diodes D1 and D2 which limit current in either direction to 300 μA within the voltage rating of the diodes (100 volts). Neons B1 and B2 ensure that these diodes are not subjected to voltages that exceed their voltage rating. R1 and R2 provide some current limiting to protect B1 and B2 should high-amplitude, high-power-capability signals be applied to either the input or to the output. Potentiometer R3 allows the dynamic impedance of two such adapters to be balanced to achieve optimum common mode rejection ratio in a differential amplifier.



NOTE: TWO OF THESE ADAPTERS ARE REQUIRED FOR USE WITH DIFFERENTIAL AMPLIFIERS.

BANDWIDTH - DC to 500kHz
 MAXIMUM SIGNAL INPUT - 50V PEAK
 ADJUST R3 FOR COMMON MODE BALANCE
 WHEN USING 2 ADAPTERS WITH A DIFFERENTIAL AMPLIFIER. CMRR OF 3A9 WITH 2 ADAPTERS IS 100,000:1.

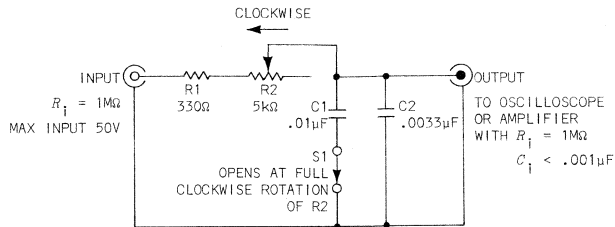
CKT NO	DESCRIPTION	TEKTRONIX PN
B1	NEON BULB 5AH-B	150-0067-00
B2	NEON BULB 5AH-B	150-0067-00
D1	FET DIODE	152-0328-00
D2	FET DIODE	152-0328-00
R1	RESISTOR 2kΩ 1/2W	323-0222-00
R2	RESISTOR 2kΩ 1/2W	323-0222-00
R3	POTENTIOMETER 500Ω	311-0634-00
--	GROMMET FOR R3	348-0003-00
--	ACCESSORY HOUSING	011-0081-00



TWO ADAPTERS IN USE WITH A TEKTRONIX TYPE 3A9 DIFFERENTIAL AMPLIFIER

LIMITS CURRENT FLOW TO THE SUBJECT FROM THE AMPLIFIER TO LESS THAN 300 μA

Fig. 29-12. A current-limiting adapter for protection from electric shock.



R2 VARIES 3dB CUTOFF FREQUENCY BETWEEN 2.3kHz AND 35kHz. WITH R2 IN FULL CLOCKWISE POSITION, S1 IS OPEN AND 3dB CUTOFF IS ABOVE 100kHz.

CKT NO	DESCRIPTION	TEKTRONIX PN
C1	.01μF 50V ±10% CAPACITOR	283-0155-00
C2	.0033μF 100V ±5% CAPACITOR	283-0051-00
R1	330Ω 1/4W ±5% RESISTOR	315-0331-00
R2	5kΩ POTENTIOMETER	311-0656-00
S1	SWITCH ON R2	--
MECHANICAL		
QTY		
1	ACCESSORY HOUSING	011-0081-00
1	KNOB - FOR R2	366-0189-00
1	NUT - FOR R2	210-0583-00
1	WASHER-FLAT - FOR R2	210-0905-00
1	WASHER-LOCKING - FOR R2	210-0046-00

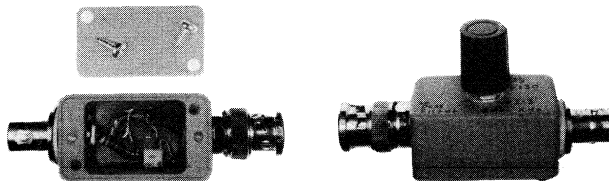
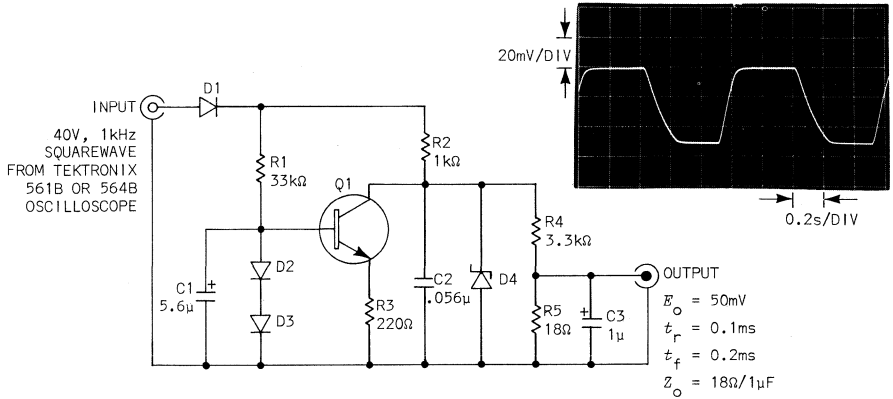


Fig. 29-13. A low-pass filter for physiological signal processing.

29.13 A LO-PASS FILTER FOR PHYSIOLOGICAL SIGNAL PROCESSING

Refer to Fig. 29-13. This filter consists of a single RC network to allow the frequency response of instrumentation to be limited in order to correspondingly limit noise. When used in conjunction with a conventional oscilloscope vertical amplifier, and with R2 fully clockwise, the input impedance is 1 MΩ in parallel with .0033 μF providing a frequency response in excess of 100 kHz. As R2 is rotated counterclockwise, the frequency response is limited between 35 kHz and 2.3 kHz depending on the position of R2.



CKT NO	DESCRIPTION	TEKTRONIX PN
C1	5.6μF TANTALUM CAPACITOR	290-0247-00
C2	.056μF ±5% CAPACITOR	285-0684-00
C3	1μF ±10% TANTALUM CAPACITOR	290-0183-00
D1	SILICON DIODE	152-0185-00
D2	SILICON DIODE	152-0185-00
D3	SILICON DIODE	152-0185-00
D4	10V ±5% ZENER DIODE	152-0149-00
Q1	2N3904 TRANSISTOR	151-0190-01
R1	33kΩ 5% 1/4W RESISTOR	315-0333-00
R2	1kΩ 5% 1/4W RESISTOR	315-0102-00
R3	220Ω 5% 1/4W RESISTOR	315-0221-00
R4	3.3kΩ 5% 1/4W RESISTOR	315-0332-00
R5	18Ω 5% 1/4W RESISTOR	315-0180-00
MECHANICAL		
QTY		
1	BOX - TERMINAL	202-0054-00
1	CONNECTOR - BNC MALE	131-0428-00
1	CONNECTOR - BNC FEMALE	131-0106-00
1	COVER - BOTTOM	200-0252-00
2	SETSCREW, 4-40	213-0048-00
2	SCREW, THREAD-FORMING	213-0141-00

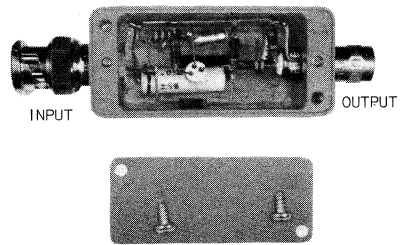


Fig. 29-14. A pulse-shaping circuit simulating the "action potential."

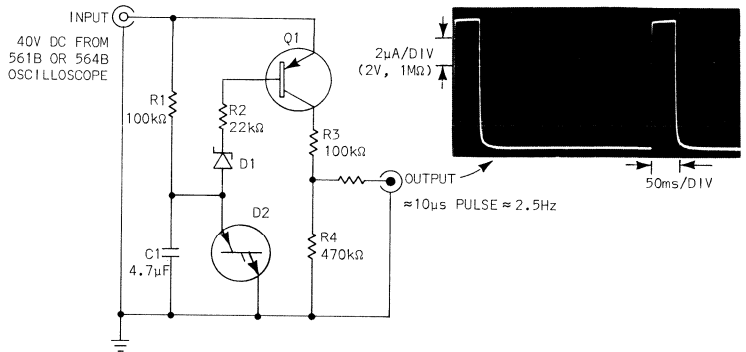
29.14 A PULSE SHAPING CIRCUIT SIMULATING THE ACTION POTENTIAL

Refer to Fig. 29-14. This device provides pulse shaping on the 1 kHz calibrating squarewave produced by Tektronix 561B or 564B Oscilloscopes. The output amplitude is 50 mV, the output risetime 0.1 ms and the output falltime 0.2 ms. The energy of the output pulse is almost all contained in the frequency spectrum from DC to 10 kHz.

When the signal produced by the oscilloscope calibrator is at 40 volts, C2 is charged in a pseudo-constant current mode via R2, providing a risetime of 0.1 ms. When the charge on C2 reaches 10 volts, D4 conducts limiting further voltage increase. This 10 volt level is attenuated by R4 and R5 to provide a 50 mV output. When the signal produced by the oscilloscope calibrator is at zero volts, a charge is maintained on C1, allowing Q1 to conduct in a pseudo-constant current mode to discharge C2 at a constant rate, providing a falltime of 0.2 ms.

29.15 A CONSTANT CURRENT PULSE SOURCE FOR GSR AND OTHER USES

Refer to Fig. 29-15. This pulse generator requires a 40 volt DC power supply which may be obtained from a Tektronix 561B or 564B Oscilloscope calibrator. Shockley diode D2 is functioning as an oscillator, the frequency of oscillation being determined by R1 and C1. The oscillations produced at the anode of D2 turn Q1 on and off via D1 and R2, thus providing a squarewave at the collector of Q1 which is attenuated via R3 and R4. A constant current is produced by this attenuated squarewave via R5.



CKT NO	DESCRIPTION	TEKTRONIX PN
C1	4.7μF TANTALUM CAPACITOR	290-0187-00
D1	33V ZENER DIODE	152-0241-00
D2	SHOCKLEY DIODE	152-0136-00
Q1	2N4250 TRANSISTOR	151-0219-00
R1	100kΩ 1/4W RESISTOR	315-0104-00
R2	22kΩ 1/4W RESISTOR	315-0223-00
R3	100kΩ 1/4W RESISTOR	315-0104-00
R4	470kΩ 1/4W RESISTOR	315-0474-00
R5	2.2MΩ 1/4W RESISTOR	315-0225-00
--	ACCESSORY HOUSING	011-0081-00

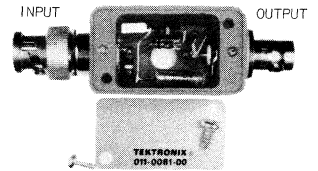


Fig. 29-15. A constant-current, pulse source for GSR and other uses.

30

DEFINITIONS

This chapter provides definitions for the biomedical terminology used throughout this book. These definitions have been taken from either *Webster's International Dictionary* (Second Edition Unabridged 1959), *Webster's Seventh New Collegiate Dictionary* (1967) or *Dorland's Illustrated Medical Dictionary* (24th Edition, 1965). The appropriate reference is given at the end of each definition. The Websters' are published by G. & C. Merriam Company, Springfield, Massachusetts, USA and the Dorland's is published by W.B. Saunders Company, Philadelphia, Pennsylvania, USA and London, U.K.

A

accretion - The process of growth or enlargement.
(Webster)

alveolus - An air cell of the lungs. (Webster)

amnion - A thin membrane forming a closed sac surrounding the embryos of reptiles, birds and mammals. It contains a thin, watery fluid, the amniotic fluid, in which the embryo is immersed.
(Webster)

amniotic - See amnion.

angstrom - Abbreviated Å. One ten-billionth of a meter. $1 \text{ Å} = 10^{-8} \text{ cm}$. (Webster)

anterior - Situated before or toward the front.
(Webster)

aorta - The great trunk artery that carries blood from the heart to be distributed by branch arteries through the body. (Webster)

aortic - Pertaining to the aorta. (Webster)

arborizations - Formation of or into a form resembling a tree in properties, growth, structure, or appearance, or such a form and arrangement. (Webster)

arrhythmia - An alteration in rhythm of the heartbeat either in time or force. (Webster)

arteriole - One of the small terminal twigs of an artery that ends in capillaries. (Webster)

artifacts - Any artificial product; any structure or feature that is not natural. (Dorland)

atria - Plural of atrium.

atrio-ventricular - Located between an atrium and ventricle of the heart. (Webster)

atrium - An anatomical cavity or passage; especially the main chamber of an auricle of the heart or the entire auricle. (Webster)

auricle - The chamber or either of the chambers of the heart that receives blood from the veins and forces it into the ventricle or ventricles. (Webster)

autonomic - Acting independently of volition; relating to, affecting, or controlled by the autonomic nervous system. (Webster)

axon - A usually long and single nerve-cell process that, as a rule, conducts impulses away from the cell body. (Webster)

B

bioelectric - See bioelectricity.

bioelectricity - The electrical phenomena which appear in living tissues. (Dorland)

biophysical - Pertaining to the branch of knowledge concerned with the application of physical principles and methods to biological problems. (Webster)

brachial - Relating to the arm or a comparable process. (Webster)

bradycardia - A slow heart rate. (Webster)

bronchi - Plural of bronchus.

bronchus - Either of two primary divisions of the trachea that lead respectively into the right and the left lung; broadly - bronchial tube. (Webster)

bundle of His - A small band of cardiac muscle fibers transmitting the wave of depolarization from the auricles to the ventricles during cardiac contraction.

C *capillaries* - Any of the smallest vessels of the blood-vascular system connecting arterioles with venules and forming networks throughout the body. (Webster)

cardiac - Pertaining to the heart. (Dorland)

cardiology - The study of the heart and its action and diseases. (Webster)

cardiovascular - Relating to the heart and blood vessels. (Webster)

catheter - A tubular medical device for inserting into canals, vessels, passageways, or body cavities usually to permit injection or withdrawal of fluids or to keep a passage open. (Webster)

cell - A small usually microscopic mass of protoplasm bounded externally by a semipermeable membrane, usually including one or more nuclei and various nonliving products, capable alone or interacting with other cells of performing all the fundamental functions of life, and forming the least structural aggregate of living matter capable of functioning as an independent unit. (Webster)

cephalic - Directed towards or situated on or in or near the head. (Webster)

cerebellum - A large dorsally projecting part of the brain, especially concerned with the coordination of muscles and the maintenance of bodily equilibrium. (Webster)

cerebrum - An enlarged anterior or upper part of the brain. (Webster)

chronaxy - The time required for the excitation of a nervous element by a definite stimulus; the minimum time at which a current just double the rheobase will excite contraction. (Dorland)

cochlea - A division of the labyrinth of the ear of higher vertebrates that is usually coiled like a snail shell and is the seat of the hearing organ. (Webster)

cornea - The transparent part of the coat of the eyeball that covers the iris and pupil and admits light to the interior. (Webster)

cortex - The outer or superficial part of an organ or body structure; especially the outer layer of gray matter of the cerebrum and cerebellum. (Webster)

cortical - Of, relating to, or consisting of the cortex. (Webster)

cranium - The part that encloses the brain. (Webster)

curare - A dried aqueous extract especially of a vine used in medicine to produce muscular relaxation. (Webster)

cytoplasm - The protoplasm of a cell exclusive of that of the nucleus. (Dorland)

D

defibrillation - The stoppage of fibrillation of the heart. (Dorland)

defibrillator - An apparatus used to counteract fibrillation (very rapid irregular contractions of the muscle fibers of the heart) by application of electric impulses to the heart. (Dorland)

dendrite - Any of the usual branching protoplasmic processes that conduct impulses toward the body of a nerve cell. (Webster)

depolarize - To cause to become partially or wholly unpolarized. (Webster)

diastole - A rhythmically recurrent expansion especially the dilatation of the cavities of the heart during which they fill with blood. (Webster)

diastolic - See diastole.

dicrotic - Having a double beat; being or relating to the second expansion of the artery that occurs during the diastole of the heart. (Webster)

dorsal - Relating to, situated near or on the back. Especially of an animal or of one of its parts. (Webster)

E *ECG* - Abbreviation for electrocardiogram. (Dorland)

ectopic - Located away from normal position. (Dorland)

EEG - Abbreviation for electroencephalogram. (Dorland)

electrocardiogram - The tracing made by an electrocardiograph. (Webster)

electrocardiograph - See electrocardiography.

electrocardiography - The making of graphic records of the electric currents emanating from heart muscle, as a method for studying the action of the heart muscle. (Dorland)

electrode - A conductor used to establish electrical current with a nonmetallic part of a circuit. (Webster)

electrodermal - See electrodermography.

electrodermography - The recording of the electrical resistance of the skin, which varies with the amount of sweating, and constitutes a sensitive index to the activity of the autonomic nervous system. (Dorland)

electroencephalogram - The tracing of brain waves made by an electroencephalograph. (Webster)

electroencephalograph - See electroencephalography.

- electroencephalography* - Recording of electric currents developed in the brain by means of electrodes applied to the scalp, to the surface of the brain, or placed within the substance of the brain. (Dorland)
- electrogastrogram* - The graphic record obtained by the synchronous recording of the electrical and mechanical activity of the stomach. (Dorland)
- electrolyte* - A nonmetallic electric conductor in which current is carried by the movement of ions. (Webster)
- electromyogram* - The tracing of muscular action potentials by an electromyograph.
- electromyograph* - See electromyography.
- electromyography* - The recording of the changes in electric potential of muscle. (Dorland)
- electrophysiology* - The science of physiology in its relations to electricity; the study of the electric reactions of the body in health. (Dorland)
- embolus* - An abnormal particle (as an air bubble) circulating in the blood. (Webster)
- embryo* - A human or animal offspring prior to emergence from the womb or egg; hence, a beginning or undeveloped stage of anything. (Webster)
- EMG* - Abbreviation for electromyography. (Dorland)
- epilepsy* - Any of various disorders marked by disturbed electrical rhythms of the central nervous system and typically manifested by convulsive attacks usually with clouding or consciousness. (Webster)
- extracellular* - Situated or occurring outside a cell or the cells of the body. (Webster)
- extracorporeal* - Situated or occurring outside the body. (Dorland)
- fibula* - The outer and smaller of the two bones of the leg. (Dorland)

fibular - Pertaining to the fibula. (Dorland)

fluoroscopy - Process of using an instrument to observe the internal structure of an opaque object (as the living body) by means of X-rays. (Webster)

frontal plane - Section drawing, etc. parallel to the main axis of the body, and at right angles to the sagittal plane. (Webster)

G
galvanic - Of, relating to, or producing a direct current of electricity. (Webster)

H
hemisphere - Half of any spherical or roughly spherical structure or organ, as demarcated by dividing it into approximately equal portions. (Dorland)

homogeneity - The quality or state of being homogeneous. (Webster)

homogeneous - Of uniform structure or composition throughout. (Webster)

I
infarct - An area of necrosis in a tissue or organ resulting from obstruction of the local circulation by a thrombus or embolus. (Webster)

infarction - See infarct.

inhomogeneities - See inhomogeneity.

inhomogeneity - Something which is not homogeneous. (Webster)

intracellular - Being or occurring within a protoplasmic cell. (Webster)

ion - An atom or group of atoms that carries a positive or negative electric charge as a result of having lost or gained one or more electrons. (Webster)

iris - The circular pigmented membrane behind the cornea of the eye. (Dorland)

iso-electric - Uniformly electric throughout, or having the same electric potential, and hence giving off no current. (Dorland)

isothermal - See isothermic.

isothermic - Having the same temperature. (Dorland)

isotropic - Exhibiting properties with the same values when measured along axes in all directions. (Webster)

isotropy - See isotropic.

L

latency - A state of seeming inactivity, such as that occurring between the instant of stimulation and the beginning of response. (Dorland)

lobe - A somewhat rounded projection or division of a bodily organ or part. (Webster)

lumen - (1) the cavity of a tubular organ (the lumen of a blood vessel). (2) the bore of a tube (as of an organ). (Webster)

M

manometer - An instrument for measuring the pressure of gases and vapors: pressure gauge. (Webster)

membrane - A thin layer of tissue which covers a surface or divides a space or organ. (Dorland)

metabolism - The sum of all the physical and chemical processes by which living organized substance is produced and maintained. (Dorland)

micron - A unit of length to one thousandth of a millimeter. $1\mu = 10^{-3}$ mm = 10^{-6} meters. (Webster)

mitochondria - Small granules or rod-shaped structures found in differential staining in the cytoplasm of cells. (Dorland)

mitral stenosis - A narrowing of the left atrio-ventricular orifice. (Dorland)

myocardial - See myocardium.

myocardium - The middle muscular layer of the heart wall. (Webster)

myograph - An apparatus for recording the effects of a muscular contraction. (Dorland)

myographic - See myograph.

- N**
- necrosis* - Death of tissue, usually as individual cells, groups of cells, or in small localized areas. (Dorland)
- neuron* - A nerve cell with its processes, collaterals, and terminations regarded as a structural unit of the nervous system. (Dorland)
- neuronal* - See neuron.
- nomography* - A graphic method by which the relation between any number of variables may be represented graphically on a plane surface, such as a piece of paper. (Dorland)
- nuclei* - Plural of nucleus. (Dorland)
- nucleus* - A central point, group, or mass about which gathering, concentration or accretion takes place -- essential portion of a cell -- group of nerve cells in the central nervous system. (Webster)
- O**
- occipital* - See occiput.
- occiput* - Of or relating to the back part of the head or skull. (Dorland)
- organ* - A differentiated structure consisting of cells and tissues and performing some specific function. (Webster)
- orthogonal* - At right angles to.
- orthogonality* - See orthogonal.
- P**
- parietal* - Of, relating to, or forming the upper posterior wall of the head. (Webster)
- permeable* - See permeate.
- permeate* - To pass through the pores or interstices. (Webster)
- peroneal* - Pertaining to the fibula or to the outer side of the leg. (Dorland)
- piezoelectric* - See piezoelectricity.

- piezoelectricity* - Electricity or electric polarity due to pressure especially in a crystalline substance. (Webster)
- plethysmography* - The recording of the changes in the size of a part as modified by the circulation of the blood in it. (Dorland)
- plastid* - Any specialized organ of the cell other than the nucleus. (Dorland)
- pneumatic* - Relating to, or using air, wind, or other gas (a) moved or worked by air pressure (b) adapted for holding or inflated with compressed air. (Webster)
- pneumograph* - An instrument for recording the thoracic movements or volume change during respiration. (Webster)
- pneumotachygraph* - An instrument for recording the velocity of the respired air. (Dorland)
- posterior* - The hinder parts of the body. (Webster)
- protoplasm* - The colloidal complex of protein, other organic and inorganic substances, and water that constitutes the living nucleus, cytoplasm, plastids, and mitochondria of the cell and is regarded as the only form of matter in which the vital phenomena are manifested. (Webster)
- protoplasmic* - See protoplasm.
- psychogalvanic* - See psychogalvanometer.
- psychogalvanometer* - A galvanometer for recording the electrical agitations produced by emotional stresses. (Dorland)
- pulmonary* - Relating to, functioning like, or associated with the lungs. (Webster)
- pupil* - The contractile aperture in the iris of the eye. (Webster)

R

radical - A group of atoms that is replaceable by a single atom, that is capable of remaining unchanged during a series of reactions, or that may show a definite transitory existence in the course of a reaction. (Webster)

radioisotope - An isotope which is radioactive, produced artificially from the element or from a stable isotope of the element by the action of neutrons, protons, deuterons, or alpha particles in the chain-reacting pile or in the cyclotron. Radioisotopes are used as tracers or indicators by being added to the stable compound under observation, so that the course of the latter in the body (human or animal) can be detected and followed by the radioactivity thus added to it. The stable element so treated is said to be "labeled" or "tagged." (Dorland)

real-time spectrum analyzer - A spectrum analyzer that performs a continuous analysis of the incoming signal with the time sequence of events preserved between input and output.

retina - The sensory membrane of the eye that receives the image formed by the lens, is the immediate instrument of vision and is connected with the brain by the optic nerve. (Webster)

rheobase - The minimum potential of electric current necessary to produce stimulation. (Dorland)

S

sagittal - Of, relating to, or situated in the median plane of the body or any plane parallel thereto. (Webster)

scalp - That part of the integument of the head which normally is covered with hair. (Dorland)

selenide - A compound of selenium with an element or radical. (Webster)

selenium - A nonmetallic element relating to sulphur and tellurium and resembling them chemically. (Webster)

- semipermeable* - Partially but not freely or wholly permeable (Webster)
- sinoatrial node* - A microscopic collection of atypical cardiac muscle fibers which is responsible for initiating each cycle of cardiac contraction.
- sinus* - A cavity in the substance of the bone of the skull that usually communicates with the nostrils and contains air. (Webster) Also, any irregular cavity, particularly in the circulatory system.
- spatial* - Pertaining to space. (Dorland)
- sphygmomanometer* - An instrument for measuring blood pressure and especially arterial blood pressure. (Webster)
- spirometer* - An instrument for measuring the air entering and leaving the lungs. (Webster)
- stereotaxic* - Pertaining to or characterized by precise positioning in space. (Dorland)
- stimulate* - To excite to functional activity. (Dorland)
- stroboscope* - An instrument for determining speeds of rotation or frequencies of vibration made in the form of a rapidly flashing light that illuminates an object intermittently. (Webster)
- stroboscopic* - Of, utilizing, or relating to a stroboscope. (Webster)
- synapse* - The point at which a nervous impulse passes from one neuron to another. (Webster)
- synaptic* - See synapse.
- systemic* - Pertaining to or affecting the body as a whole. (Dorland)
- systole* - The contraction, or period of contraction, of the heart, especially that of the ventricles. It coincides with the interval between the first and second heart sound, during which blood is forced into the aorta and the pulmonary trunk. (Dorland)

systolic - See systole.

T

tachycardia - Relatively rapid heart action.
(Webster)

tetrahedron - A solid body having four faces.
(Webster)

thermistor - An electrical resistor made of a material whose resistance varies sharply in a known manner with the temperature. (Webster)

thermocouple - A thermoelectric couple (a union of two conductors -- as bars of dissimilar metals joined at their extremities -- for producing a thermoelectric current) used to measure temperature differences. (Webster)

thermoelectric - Of or relating to phenomena involving relations between the temperature and the electrical condition in a metal or in contacting metals. (Webster)

thoracic - See thorax.

thorax - The part of the body of man and other mammals between the neck and the abdomen.
(Webster)

thrombus - A clot of blood formed within a blood vessel and remaining attached to its place of origin. (Webster)

tibia - The inner and usually larger of the two bones of the vertebrate hind limb between the knee and ankle. (Webster)

tibial - See tibia.

tissue - An aggregation of similarly specialized cells united in the performance of a particular function. (Dorland)

torso - The human trunk. (Webster)

trachea - The main trunk of the system of tubes by which air passes to and from the lungs.
(Webster)

transducer - A device that is actuated by power from one system and supplies power in any other form to a second system. (Webster)

U

ulnar - Pertaining to the inner and larger bone of the forearm, on the side opposite that of the thumb. (Dorland)

utero - (1) A combining form from uterus, (Webster), as *utero-vaginal*.
(2) The Latin dative of uterus, as *in utero*, in the uterus.

uterus - The hollow muscular organ in female animals which is the abode and the place of nourishment of the embryo and fetus. (Dorland)

V

vasoconstriction - Narrowing of the lumen of blood vessels especially as a result of vasomotor action. (Webster)

vasomotor - Any element or agent that affects the caliber of a blood vessel. (Dorland)

vector - A quantity that has magnitude, direction, and sense and that is commonly represented by a directed line segment whose length represents the magnitude and whose orientation in space represents the direction. (Webster)

ventricle - A chamber of the heart which receives blood from a corresponding atrium and from which blood is forced into the arteries. (Webster)

ventricular - See ventricle.

venule - A small vein; especially one of the minute veins connecting the capillary bed with the larger systemic veins. (Webster)

vertex - The top of the head. (Webster)

viable - Capable of living; especially born alive, with such form and development of organs as to be normally capable of living. (Webster)

viability - See viable.

INDEX

- Acceleration transducer,
288-290
- Acoustical techniques, 201-211
- Acoustic coupler, 434-435
- Action potential (defined),
14-15
evoked, 148-170
simulated, 321
- Adioventricular rhythm
(defined), 66
- Alpha rhythm, 44, 45, 131,
133, 147
- Amplifiers, 63-65, 156-158,
175-176, 293-317
differential, 63-65, 103,
156-158, 175-176, 181,
196, 226, 294-303,
305-335
operational, 124, 156-158,
326-335
single-ended, 304
- Analog to digital conversion,
432, 435
- Anions (defined), 8
- Aorta, 25-27, 95, 112, 475
- Aortic valve, 26-27
- Arrhythmia (defined), 66, 476
- Arterial catheterization, 95,
439
- Arterioles (defined), 25, 476
- Artifact (defined), 133, 476
- Atrioventricular bundle, *see*
Bundle of His
- Atrioventricular conduction
time, 30
- Atrioventricular node, 27, 29
- Atrioventricular valves,
26-27, 207-209, 217
- Atrium, 25-30, 95, 476
- Augmented vector (defined),
54
- Auricle, 66, 476
- Autonomic nervous system,
195-199, 217
- Aveoli (defined), 115
- Average reference recording
(EEG), 139-140
- aVF, aVL, aVR (defined), 54
- AV node, *see* Atrioventricular
node
- Axial vectorcardiogram, 74-75
- Axon (defined), 39, 476
- Ballistocardiogram, 218
- Bandwidth (amplifiers), 319,
321-325, 333
- Bandwidth limiting, 321-325,
333
- Baseline drift, *see* Drift
- Basilar membrane, 37
- Beta rhythm, 133
- Bioelectric potential (defined),
7
- Biphasic stimulation, 172-173
- Bipolar ECG, 53-54
- Bipolar recording (EEG),
139-140
- Blondheim configuration, 83
- Blood flow measurements, 93,
108-112, 217-218, 430-431
dye-dilution techniques,
112-113, 217
electroturbinometer, 112
217
flowmeters, 108-111, 217
isothermal, 112, 217
thermistors, 112, 217
ultrasonic, 112, 211, 217
- Blood pressure measurements,
93-107, 217-218, 259, 290,
387, 439, 442-445
brachial artery, 95
catheter use, 94-97, 439
direct, 93-99
femoral artery, 97

- Blood pressure measurements
 - (continued)
 - indirect, 93, 101-107
 - instrumentation, 102-107, 290, 387
 - Korotkoff sounds, 101-102, 439
 - needle use, 97
 - plethysmograph, 104-107, 217, 439
 - pulse sensor, 104-107
 - relative, 93, 104-107, 217, 439, 442-445
 - sphygmomanometer, 100, 217
 - syringe use, 97
 - terminology, 98-99
 - dye-dilution techniques, 112-113, 217
 - electroturbinometer, 112, 217
 - flowmeters, 108-111, 217
 - isothermal, 112, 217
 - thermistors, 112, 217
 - ultrasonic, 112, 211, 217
- Blood volume measurements, 113, 217-218
- Body temperature, *see* Temperature
- Bonded strain gage, 284-285
- Bootstrapping, *see* Input guarding
- Brachial artery, 95
- Brain, 38-45
- Bronchi, 114-115, 477
- Bundle of His, 27, 29-30, 66, 477
- Burger triangle, 52
- Cameras, 78, 87, 88, 389-397
- Cardiac catheters, *see* Catheters
- Cardiac output (defined), 112
- Cardiac vector (defined), 50
- Cardiography, 209
- Cardiovascular system, 23-30, 49-112, 207-208, 437-446
 - monitoring in intensive care, 437-446
- Cardioversion, 267
- Catheters, 63, 94-97, 112, 258-261, 439, 477
 - cardiac, 94-97, 258-261
 - electrode, 63
- Cations (defined), 8
- Cautery, 240-252-253, 259, 265-266
- Cell action potential, 7, 14-21
- Cell electrical activity, 7-21, 42-43
- Cell size, 7
- Cell stimulation, 16-18
- Central sulcus, 41
- Cerebellum, 41, 477
- Cerebral circulation, 25
- Cerebral hemisphere, 41-42, 133
- Cerebrum, 41, 478
- Character generator, 387
- Chart recorders, *see* Graphic recorders
- Chemical analysis transducers, 292
- Chemical gradient (defined), 8
- Chopped blanking, 352, 359
- Chronaxy (defined), 178, 478
- Closed-circuit television, 440-441, 443-445
- CMRR, *see* Common-mode rejection ratio
- Common-mode rejection ratio (defined), 295-296
- Common-mode signal (defined), 295
- Computers, 159-163, 326, 381, 387, 430-436, 446
 - analog, 430-431
 - digital, 432-433, 446
 - signal averaging, 159-163, 376, 435
 - software, 435
 - terminals for, 434-435
 - time sharing, 432-435
- Contact lens electrode, 169, 217
- Continuous-motion photography, 389, 394-397, 399

- Corpus callosum, 41-42
- Cortex, 40-42, 45, 131, 134, 145, 164-166, 478
- Cortical stimulation, 364
- Cranium, 41, 478
- Cube vectorcardiogram, 74-75
- Current limiting, 261-262, 470
- Curvilinear graphic recordings, 405
- Cyst detection, 210-211
- Damping (graphic recorders), 403
- Data processing, 423, 430-437
- Data transmission, 423-430
- Defibrillation, 240, 266-267, 309, 373-374, 478
- Delta rhythm, 45, 133, 146
- Dendrite (defined), 39, 478
- Densitometer, 112
- Depolarization (defined), 12
 - asynchronous, 21
 - synchronous, 18
 - threshold, 15, 43
 - travelling wave of, 21
- Dermometer, 199
- Diaphasic response (defined), 187
- Diastole (defined), 26, 479
- Diastolic pressure (defined), 98
- Dicrotic notch (defined), 99
- Differential amplifier, 63, 65, 103, 156-158, 175-176, 181, 196, 226, 294-303, 305-310
- Differential capacitor transducer, 288
- Differential transformer transducer, 287
- Digital indicators, 387
- Digital-to-analog conversion, 432
- Displacement transducers, 120-121, 283-289
- Display devices, 77-78, 338, 356-357, 375-386
- Doppler ultrasound, 211
- Drift (amplifier), 225-231, 246-247, 310, 315-317
- Drift catheters, 97
- Dual beam (defined), 342
- Dual trace (defined), 342
- Dye dilution techniques, 112-113
- Dynamic window (defined), 310
- Ear, 36-37
- ECG, *see* Electrocardiography
- Echoencephalography, 205
- Ectopic beats (defined), 66
- EDR, *see* Electrodermal response
- EEG, *see* Electroencephalography
- Einthoven's triangle, 50-59
- EKG, *see* Electrocardiography
- Elastic force gage, 121, 217
- Electrical skin resistance, 195, 199
- Electric shock, 255-267
- Electric shutter, 392
- Electrocardiography, 28-30, 49-91, 217-218, 231, 237, 240, 249-253, 259-261, 266-267, 295, 298, 317, 385, 405, 426, 432, 436, 438, 440, 442-446, 450-451, 479
- Einthoven's triangle, 50-59
 - fetal, 80-91, 450-451
 - frontal plane, 48-59, 66
 - interpretation of, 65-66
 - sagittal plane, 48-49, 63
 - transverse plane, 48-49, 60-62
- vectorcardiography, 67-79
- Electrocautery, 240, 252-253, 259, 265-266
- Electrode adapters, 237
- Electrode application, 245
- Electrode cables, 237
- Electrode double layer (defined), 223
- Electrode materials,
 - copper, 219, 226-227, 229, 237
 - lead, 198-199, 235

- Electrode materials (continued)
- silver, 135, 198-199, 234-235, 237, 242-243, 245, 247
 - silver/silver chloride, 135, 169, 173, 224, 226-229, 231-232, 237, 240, 247
 - stainless steel, 226-227, 229, 240, 242
 - zinc/zinc sulfate, 224
- Electrode offset potential (defined), 221
- Electrode paste, 225, 233
- Electrode placement, 52-63, 69-74, 82-87, 135-136, 143, 169
- ECG, 52-63
 - EEG, 135-136
 - ERG, 169
 - fetal ECG, 82-87
 - intracranial, 143
 - VCG, 69-74
- Electrode potentials, 220
- Electrode types, 15, 34-35, 63, 98, 135, 143, 150-156, 164, 166, 169, 173, 182, 187, 189, 198-199, 217, 219-248, 258, 261
- catheter, 63, 258, 261
 - contact lens, 169, 217
 - micro, 15, 34-35, 98, 150-156, 164, 166, 217, 243-245
 - glass, 151-153, 243-245
 - metal, 151, 243-245
 - pressure, 98
 - needle, 34-35, 143, 166, 182, 189, 217, 240-242
 - concentric, 35, 143, 166, 187, 242
 - insulated, 35, 143, 242
 - spray on, 240
 - surface, 135, 169, 173, 189, 198-199, 217, 219-240, 369, 439
 - direct contact, 233
 - disposable, 240
 - indirect contact (floating), 233
 - reusable, 233-239
- Electroencephalography, 44-45, 131-147, 162-163, 217-218, 231, 235, 237, 245, 249, 254-255, 295, 317, 406, 432, 480
- electrodes, 134-139, 231, 235, 237, 245
 - external stimuli, 147
 - line interference, 135, 254-255
 - recording modes, 139-140
 - signal averaging, 162-163
 - spectrum analysis, 141
- Electrogastrogram, 193, 217-218, 480
- Electromagnetic flowmeter, 108-111, 217
- Electromyography, 35, 166, 181-193, 217-218, 242, 249, 367, 480
- Electronystagmogram, 218
- Electroretinogram, 37, 169-170, 217-218
- Electroturbinometer, 112, 217
- Emergency gate, 34
- EMG, *see* Electromyography
- Endosomatic response, 195
- Epilepsy, 145-146, 480
- Equivalent noise resistance, 311
- ERG, *see* Electroretinogram
- ESR, *see* Electrical skin resistance
- Evoked action potentials, 45, 149-170
- Excess noise (amplifier), 315, 321-322
- Expiratory reserve volume (lung), 116
- Eye, 36-37, 206, 210-211
- Fault current detector, 264
- F-EGG, *see* Fetal electrocardiography
- Femoral artery, 97

- Fere effect, 195
- Fetal electrocardiography,
80-91, 217
conflict with maternal ECG,
82, 90
electrode placement, 82-87
multiple pregnancy, 90
subject preparation, 82-83
- Fetal ultrasonography, 210-211
- Flicker noise, 315
- Flowmeter, 108-111, 217
- Force transducers, 284-288,
290
- Frank electrode system, 70-73
- Frequency response, 319, 404,
416-418, 471
amplifiers, 319
graphic recorders, 404
tape recorders, 416-418
- Frontal lobe, 41
- Frontal plane ECG, 50-59,
66-67
bipolar limb leads, 53-54
Burger's triangle, 52
cardiac vector, 50-57
chest cluster, 58
Einthoven's triangle, 50-59
electrode position, 53-58
noise, 58
unipolar limb leads, 53-59
waveform polarity, 54
- Frontal VGG, 68-77
- Functional residual capacity
(lung), 116
- Gage factor (defined),
272-273
- Galvanic skin reflex,
194-198, 217-218, 235,
237, 474
- Galvanometric graphic
recorders, 403-404
- Gamma rhythm, 133
- Geiger-Muller counter, 292
- Graphic recorders, 63-64,
129, 133, 136, 140,
399-411, 444-445
- Grounding, 249-255, 259-260,
262-266
- Ground loop, 249-251, 425
- GSR, *see* Galvanic skin reflex
- Hair cells (ear), 37
- Half-cell potential (defined),
219
- Heart potentials, 23-30, 49-91,
217, 249-267 *see also*
Electrocardiogram
fetal electrocardiogram,
80-91
safety, 249-267
vectorcardiogram, 67-79
- Hepatic portal circulation, 25
- Hodgkin-Huxley theory, 7-8
- Horizontal amplifiers
(oscilloscope), 339
- H reflex, 187-189, 217
- Hum (defined), 310
- Impedance pneumograph,
121-122, 217
- Indicators, 375, 387-388
- Indifferent electrode
(defined), 54
- Inferior vena cava, *see*
Vena cava
- Input guarding (defined),
302-303
- Input neutralization, 157-158,
458-459
- Input normalizers, 359
- Inspiratory capacity (lung),
116
- Inspiratory reserve volume
(lung), 116
- Intensive care, 437-446
- Iontophoresis, 247
- Jitter, 162
- Johnson noise, 315, 321-322
- Korotkoff sounds, 101-102, 287,
439
- Latency (defined), 164, 482
- Light-sensitive resistor
transducer, 292
- Line interference, 135, 137,
254-255, 294-295, 325
- Line-pair (defined), 383
- Lowpass filter, 158
- Lumen (defined), 109, 482

- Lung, 115
 capacity of, 116-117
 physiology of, 114-115
- Magnetic tape, *see*
 Tape recording
- Manometer, 98, 217, 482
- Median nerve, 192
- Medulla oblongata, 41
- Microelectrode, *see*
 Electrode types
- Micromanipulator, 166
- Middle ear, 36-37
- Minute respiratory volume
 (defined), 125
- Mitral valve, *see*
 Atrioventricular valve
- Modem, 434-435
- Monophasic response (defined),
 187
- Motoneuron, 31
- Motor cortex, 41-42
- Motor end-plate, 31
- Motor nerve propagation
 velocity, 189-190
- Motor unit (defined), 31
- Moving-coil meter, 375, 387
- M reflex, 188-189
- Multiplexing (defined), 429
- Muscle action, 32, 180-193
- Myelin sheath, 38
- Myocardial infarction, 23,
 437
- Myocardium, 30, 482
- Myography, 180, 217, *also see*
 Electromyography
- Needle electrodes, *see*
 Electrode types
- Nernst relation, 8, 11
- Nerve conduction, 189-192
- Nerve propagation velocity,
 189-192
- Nerve tissue threshold
 (defined), 177
- Net gradient (defined), 11
- Neuroglia (defined), 40
- Node of Ranvier, 38
- Noise, 135, 158-163, 225,
 227, 246-247, 310-315,
 421, 471
- Nucleus (defined), 39
- Occipital lobe, 41, 131, 483
- Offset potential (defined), 221
- One over f noise, 315, 322
- Optical graphic recorders, 411
- Optical transducers, 286, 292
- Oscillographs, *see*
 Graphic recorders
- Oscilloscope cameras, 78, 82,
 88, 389-397
- Oximetry, 218
- Pacemaker, 258, 370-372, 440,
 444
- Parietal lobe, 41, 483
- Peak-to-peak noise, 313
- Peroneal nerve, 192, 483
- Persistence (phosphor), 341
- PGR, *see*
 Psychogalvanic reflex
- Phonocardiogram, 218
- Photographic film, 389,
 391-392, 397
- Photoresistor, 292
- Physiological monitors, 64-65
- Piezoelectric transducer,
 286-287
- Plethysmograph, 104-105,
 217-218, 439, 442-443,
 484
- Pneumograph, 117-122, 217,
 442-445, 483
 conductive-fluid in tube,
 121, 284
 strain gage, 121
 thermistor, 118, 442-443
 torso impedance, 122
- Pneumotachograph, *see*
 Pneumotach
- Pneumotach, 122-125, 217-218,
 484
- Polarized cell (defined), 11
- Potentiometric graphic
 recorders, 404
- Power supply noise, 317
- Pressure transducers, 93, 95,
 97-98, 288-290
- PR interval, 28
- Probes (oscilloscope),
 356-359

- Projected Graticule, 390-391
 Protection (subject), *see*
 Safety
 Psychogalvanic reflex, 195
 Pulmonary arteries, 25-27,
 112, 114
 Pulmonary circulation
 (defined), 25
 Pulmonary valve, 26-27
 Pulmonary veins, 27, 114
 Pulse generators, 361-374
 Pulse sensor, *see*
 Plethysmograph
 Purkinje system, 27, 29
 P wave, 28, 30, 57, 66, 74
 QRS segment, 28, 30, 55, 57,
 66, 72, 74, 77, 91
 QT interval, 28
 Radiation transducer, 292
 Radioisotope, 112
 Rate intensification, 348
 Rectilinear graphic
 recordings, 406
 Reflex response, 33-34
 Refractory period (defined),
 16
 Regenerative breakdown, 15
 Renal circulation, 25
 Residual volume (lung), 116
 Resistance pneumograph,
 121-122
 Resistive transducers,
 270-285, 454-455
 Resolution (CRT), 382-385
 Respiration, 114-127, 217,
 442-445
 air flow, 122-125
 instrumentation, 118-127
 lung volume, 116-117
 physiology of, 115-117
 pneumograph, 117-122, 217,
 442-445, 454-457
 pneumotach, 122, 125
 Resting potential, 7
 Retina, 36-37
 Rheobase (defined), 177, 485
 Risetime, 321
 RMS noise, 311
 R wave, 28, 50, 55, 82, 88,
 91, 267
 Safety, *facing* 1, 65, 79, 169,
 224, 255-267, 470
 Sagittal plane ECG, 48-49, 63,
 67
 Sagittal VCG, 68-77
 SA node, *see*
 Sinoatrial node
 Schwann cell, 38
 S-D curve, *see*
 Strength-duration curve
 Semiconductor strain gage,
 284-285
 Sensitivity factor (defined),
 276
 Sensory cortex, 41-42
 Sensory nerve propagation
 velocity, 190-192
 Sensory system, 36-37
 Shielded cable, 423-425
 bandwidth, 423-425
 losses, 423
 Shock (electric), 255-267, 470
 Signal averaging, 159-163, 326
 Simulated action potential, 321
 Sinoatrial node, 27, 29-30, 486
 Skeletal muscle (defined), 193
 Skin resistance, *see* Galvanic
 skin reflex
 Slave oscilloscope, 338,
 351-360, 378
 Smooth muscle (defined), 193
 Sodium-potassium pump
 (defined), 15
 Soma (defined), 39
 Spectrum analysis (EEG), 141
 Speed Computer, 392
 Sphygmomanometer, 100, 217, 486
 Spinal nerves, 38-45
 Spirogram, 117, 124, 217
 Spirometer, 117, 126, 217, 486
 Spot size, 383
 Spray-on electrode, 240
 Sterotaxic instruments, 166-167
 Stewart-Hamilton dye dilution
 technique, 112
 Stimulating electrodes, 247

- Stimulations, 166, 169-179,
 361-374
 biphasic, 172-173
 constant current, 172-173,
 367-369
 constant voltage, 172,
 367-369
 cortical, 364
 grounded, 175, 187
 isolated, 175-176, 364,
 368-370
 light, 169
 Stimulus artifact (defined),
 176
 Stimulus isolation, 175-176,
 364, 368-370
 Stimulus threshold (defined),
 16
 Strain (defined), 273
 Strain gage, 284-285
 Strength/duration curves,
 176-179, 217
 Striated muscle (defined),
 193
 Strip chart recorder, *see*
 Graphic recorders
 Stroke, 437
 ST segment, 28
 Subject protection, *see*
 Safety
 Superior vena cava, *see*
 Vena cava
 Surface electrodes, *see*
 Electrode types
 S wave, 28, 55
 Sweat gland activity, 184-191
 Sweep blanking, 352, 359
 Sweep generators
 (oscilloscope), 339
 Sweep stepping, 385-386
 Synapse (defined), 39
 Systemic circulation
 (defined), 25
 Systole (defined), 26, 486
 Systolic pressure (defined),
 98
 Tachycardia (defined), 66, 487
 Tangential noise, 158, 313
 Tape recording, 63, 136,
 413-421, 444
 direct recording, 415-416
 frequency response, 416-418
 indirect FM recording,
 417-418
 noise, 421
 tape speeds, 415, 419
 transport mechanism, 419-420
 Tarchanoff effect, 195
 TA wave, 28, 30
 Telemetry, 423, 426
 Television, 360, 379-381
 Temperature, 128-129,
 217-218, 387, 410, 440,
 442-445
 Temporal lobe, 41
 Terminal arborization
 (defined), 39
 Tetrahedron vectorcardiogram,
 74-75
 Thermal noise, 315
 Thermistors, 112, 118-121,
 128-129, 217, 270, 275,
 291, 440, 442-433,
 454-457, 487
 blood flow, 112
 body temperature, 128-129
 pneumograph, 117-121, 217,
 442-433, 454-457
 respiration, 118-121
 temperature, 128-129
 Thermocouple, 291, 487
 Thermometer, 129, 217
 Theta rhythm, 133
 Tibial nerve, 192
 Tidal volume (lung), 116, 218
 Time-division multiplexing,
 428-430
 Time jitter, 162
 Time-motion ultrasonography,
 207-209
 Trachea, 114, 487
 Transducer systems, 269-292,
 439
 Transverse plane ECG, 48-49,
 60-62, 67
 Transverse VCG, 68-77

- Travelling wave of
 - depolarization, 21, 34-35
- Tricuspid valve, *see*
 - Atrioventricular valve
- Triphasic response (defined), 187
- Tumors, 146
- T wave, 28, 30, 57, 66, 74, 256, 267
- Ulnar nerve, 192
- Ultrasonic scanning, 209-211, 217
- Ultrasonography, 201-211, 217
- Unbonded strain gage, 284-285
- Unipolar ECG, 53-54
- Unipolar esophageal lead ECG (defined), 63
- Unipolar recording (EEG), 139-140
- U wave, 28, 30
- VCG, *see* Vectorcardiography
- Vectorcardiography, 67-79, 217
 - axial electrodes, 74-75
 - cube electrodes, 74-75
 - electrode placement, 69-74
 - Frank electrode system, 70-73, 468-469
 - frontal, 68
 - instrumentation, 70-79
 - normal response, 72, 74
 - sagittal, 68
 - spatial, 68
 - tetrahedron electrodes, 75
 - transverse, 68
- Velocity transducer, 289
- Vena cava, 25, 95, 112
- Ventricle, 25-30, 95
- Ventricular fibrillation, 66, 256-258, 267
- Vertical amplifiers (oscilloscope), 338-339
- VF, VL, VR (defined), 54
- Vibration transducer, 289
- Video display, 360, 379-381
- Visual cortex, 41-42, 164-165
- Vital capacity (lung), 116
- V lead measurements (defined), 60
- Voltage differentiation and integration, 334-335
- Waveform generators, 363-364
- Weight transducer, 288, 290
- Wheatstone bridge, 271-282, 287
- X-Y chart recorders, 411

REFERENCES TO TEKTRONIX PRODUCTS

- C-10 Trace-Recording Camera, 78, 390, 392
- C-12 Trace-Recording Camera, 390-391, 397
- C-27 Trace-Recording Camera, 390, 392
- C-30A Trace-Recording Camera, 82, 88, 390, 392
- C-50 Trace-Recording Camera, 390, 392
- O Operational Amplifier, 450
- 2A63 Differential Amplifier, 103
- 2B67 Time Base, 106, 108, 120, 124, 156, 180, 205-206, 282-283, 345, 356-358, 360
- 3A3 Dual-Trace Differential Amplifier, 344
- 3A6 Dual-Trace Amplifier, 384, 386
- 3A72 Dual-Trace Amplifier, 181, 344
- 3A74 Four-Trace Amplifier, 79, 344, 358, 360
- 3A75 Amplifier, 344
- 3A8 Operational Amplifier, 124, 156-158, 181, 326, 330-335, 344, 348-349, 458-466
- 3A9 Differential Amplifier, 181, 196, 278, 282-283, 307, 311, 317, 322-323, 344, 348, 357
- 3B4 Time Base, 181, 345, 384, 386
- 3C66 Carrier Amplifier, 98, 103, 106-107, 120-123, 180, 280-281, 344
- 3L5 Spectrum Analyzer, 142-143, 344
- 129 Plug-in Unit Power Supply, 330
- 160A, 161, 162, 163, 360 Pulse Generator Series, 364-367
- 410 Physiological Monitor, 64-65, 77-79, 85-88, 104-105, 118-119, 144, 169-170, 233-237, 261-262, 282-283, 305, 309, 346-347, 393, 409, 438, 442-443, 450-457
- 504 Oscilloscope, 160
- 561B Oscilloscope, 79, 197, 205-206, 280-281, 342, 345, 356, 360, 473-474
- 564B Storage Oscilloscope, 79, 103, 106, 120, 122, 124, 127, 142-143, 156-157, 180, 181, 197, 205-206, 280-283, 342, 345, 356-357, 384, 386, 473-474
- 565 Dual-Beam Oscilloscope, 342, 345, 348
- 601 Storage Display Unit, 356-357, 376-378, 382
- 602 Display Unit, 356-357, 376-378, 382-383
- 611 Storage Display Unit, 77-78, 376-378, 382, 384-385, 435
- T4002 Graphic Computer Terminal, 434-435
- 4501 Scan Converter Unit, 358-360, 377, 379-381
- 5031 Dual-Beam Storage Oscilloscope, 78, 338, 342-343

NOTES