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Ultrasonics in Medical Diagnosis [and Discussion]

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Ultrasonics in medical diagnosis

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Beginning with the wave equation, the characteristics of ultrasound are described in terms of propagation, reflexion, beam formation, scattering and attenuation. In medical diagnosis, ultrasound in the low megahertz frequency range is both generated and detected by piezoelectric transducers. At these frequencies the wavelength is in the order of 1 mm; this is one of the fundamental limitations of attainable resolution. Because attenuation increases with frequency, it is necessary to compromise between resolution and required penetration.

The time delays between pulse transmission and echo reception are proportional to the distances between the transducer and the reflectors along the ultrasonic beam. The echo amplitudes are controlled both by the characteristics of the reflectors, and by attenuation in the intervening media.

The pulse–echo method is used to produce A-scan, B-scan and C-scan displays. At present the B-scan is the most important. It is used for the study of structure motion by time–position recording, and for tomography by two-dimensional scanning. Grey-scale display gives clues about tissue characteristics. Manually operated, automatic, and very rapid (real-time) two-dimensional scanners are in clinical use, especially in obstetrics and gynaecology, internal medicine and cardiology.

The Doppler frequency-shift method is used to detect the movements of structures (especially the foetal heart) and to measure blood flow velocities. Analysis of Doppler blood-flow signals gives data on vessel characteristics. Pulsed Doppler systems can be used to estimate both velocity and position.

Newer methods include phase compensation for the distortion otherwise introduced by the skull in brain scanning, tissue characterization, computerized tomography, Doppler blood flow transfer function analysis and tumour detection.

1. Introduction

During the last decade, ultrasonic diagnosis has developed into an important method of routine investigation. According to the particular clinical problem, ultrasound may be complementary to other diagnostic techniques, or it may replace them. More and more doctors depend on the data obtained with ultrasound, and many patients are already demanding this method of examination.

Ultrasonic diagnosis is based on the measurement of the reflexion, scattering and absorption of megahertz frequency waves as they propagate through body tissue. Ultrasound can be used to visualize the anatomy of soft-tissue structures in real time, with resolutions of the order of 1 mm. Data can be obtained sufficiently rapidly to allow physiological movements to be studied. Ultrasonic diagnosis is generally non-invasive, and it seems to be harmless (Wells 1977*a*).

2. Physical principles

The basis of ultrasonic diagnosis has been described in detail elsewhere (Wells 1977*a*), and so only a brief outline is presented here. The wave equation for longitudinal mechanical waves is

$$\frac{\partial^2 u}{\partial z^2} = \frac{1}{c^2}\frac{\partial^2 u}{\partial t^2}, \tag{1}$$

where u is the particle displacement amplitude, z is the position in space along the direction of propagation, t is the time, and c is the propagation velocity. The velocity is related to the elasticity K and density ρ of the medium in which the wave is travelling, according to the equation

$$c = (K/\rho)^{\frac{1}{2}}. \tag{2}$$

At a plane boundary between two media with velocities c_1 and c_2 respectively,

$$\theta_{\mathrm{i}} = \theta_{\mathrm{r}} \tag{3}$$

and

$$(\sin\theta_{\mathrm{i}}/\sin\theta_{\mathrm{t}}) = c_1/c_2, \tag{4}$$

where θ_{i}, θ_{r} and θ_{t} are respectively the angles of incidence, reflexion and refraction. At normal incidence

$$I_{\mathrm{r}}/I_{\mathrm{i}} = [(Z_2-Z_1)/(Z_2+Z_1)]^2, \tag{5}$$

where I_{i} and I_{r} are respectively the intensities of the incident and reflected waves, and Z_1 and Z_2 are the characteristic impedances of the two media. The characteristic impedance Z of a medium is given by

$$Z = \rho c. \tag{6}$$

The situation to which (5) applies is called *specular reflexion*, and it implies that the reflecting boundary is both smooth and extensive in relation to the wavelength λ. By defining a quantity ψ related to the dimension of the obstacle (such as ensemble diameter), two situations can be distinguished, each with a corresponding value of scattering cross section S:

$$S = 1 \quad \text{when} \quad \psi \gg \lambda, \tag{7}$$

and

$$S \propto k^4\psi^6 \quad \text{when} \quad \psi \ll \lambda, \tag{8}$$

where $k = 2\pi f$, and the frequency $f = c/\lambda$. In this way, specular reflexion is described by (7) and Rayleigh scattering by (8). Moreover, with obstacles of intermediate size (or with rough surfaces), some energy may be scattered while some is specularly reflected.

The ultrasonic transducer used in medical diagnosis is commonly in the form of a disk of ferroelectric material (see §4). With continuous wave excitation producing ultrasound of wavelength λ,

$$I_z/I_0 = \sin^2\{(\pi/\lambda)[(a^2+z^2)^{\frac{1}{2}}-z]\}, \tag{9}$$

where I_0 is the intensity at the surface of the transducer, I_z is the intensity at a distance z from the transducer along the central axis, and a is the radius of the transducer. In the far field, beyond the last axial maximum (at $z = a^2/\lambda$, provided that $a^2 \gg \lambda^2$), the directivity function is given by

$$D_{\mathrm{s}} = \frac{2J_1(ka\sin\theta)}{ka\sin\theta}, \tag{10}$$

where θ is the angle relating D_{s} to the central axis of the beam, and J_1 is the first order Bessel function. In the near field the diffraction pattern is characterized by maxima and minima of decreasing complexity moving away from the transducer.

If the transducer produces a transient ultrasonic disturbance, the pulse has its energy spread over a spectrum of frequency, corresponding to its bandwidth. Consequently, it is not possible to assign single values to λ or k in (9) or (10). Physically the diffraction pattern is smeared, and the situation is further complicated by time contact boundary conditions (see, for example, Papadakis & Fowler 1971; Beaver 1974; Robinson, Lees & Bess 1974).

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3. Ultrasonic properties of biological materials

The properties of biological materials which are of principal importance in ultrasonic diagnosis are the attenuation, the velocity, and the scattering cross section. Attenuation is due to absorption, scattering and reflexion; relaxation is the main contributing mechanism in soft tissues, and at least in the low megahertz frequency range the attenuation coefficient is roughly proportional to frequency. Dispersion does occur, but it is too small to be significant in practice. Values of attenuation and velocity are given in table 1. Data on scattering cross sections are not yet available, nor is it clear at present how this information could be presented for convenient and universal application.

Table 1. Ultrasonic properties of some materials of importance in medical diagnostics

material	velocity, c / m s^{-1}	density, ρ / kg m^{-3}	characteristic impedance, Z / 10^6 kg m^{-2} s^{-1}	attenuation coefficient, α, at 1 MHz / dB cm^{-1}	frequency dependence of α (1–5 MHz)
air	330	1.2	0.0004	10	f^2
blood	1560	1060	1.7	0.1	$f^{1.2}$
bone	2700–4100	1400–1800	3.8–7.4	3–10	$f^{1.5}$
fat	1460–1480	920	1.4	1	f
liver	1540–1580	1100	1.7	1	f
lung	650–1160	400	0.3–0.5	40	$f^{0.6}$
muscle	1545–1630	1100	1.7	1.5–2.5	f
water	1480	1000	1.5	0.002	f^2

Data collected by Wells 1977*a*.

As indicated in table 1, the velocities in soft tissues are all around 1500 m s^{-1}; this corresponds to a wavelength of 1 mm at a frequency of 1.5 MHz. In any imaging system, one of the factors which limits the attainable resolution is the wavelength of the radiation used to form the image. In medical diagnosis using ultrasound, it is desirable to visualize structures with dimensions of the order of a millimetre or less. In principle, the resolution should increase as the wavelength is reduced by increasing the frequency. Unfortunately, an increase in frequency is accompanied by an increase in attenuation, and so it is necessary to compromise at any particular range between shorter wavelength, and poorer signal:noise ratio. In practice, as discussed in §4, the maximum penetration which can be achieved without signal averaging is around 200–300 wavelengths, and so frequencies in the range 2–3 MHz are generally used for abdominal and cardiological investigations, 5–10 MHz for visualizing superficial blood vessels, and 10–20 MHz for studying the eye.

4. The pulse–echo method

(*a*) *Physical considerations*

The ultrasonic pulse–echo method depends on the measurement both of the times which elapse between the transmission of an ultrasonic pulse and the reception of its echoes, and of the amplitudes of the echoes.

Ultrasound is both generated and detected by transducers based on the piezoelectric effect. Polarized lead zirconate titanate is generally used, with a layer of intermediate characteristic impedance on the front face to improve efficiency, and a backing block to provide damping for short pulse operation.

The factors which determine the distribution of the ultrasonic field are discussed in §2. In a pulse–echo system the resolution cell is the volume of material within which the interaction providing the data takes place. The dimensions of the resolution cell depend on the range of the target, and the effective dynamic range (Wells 1969). In essence, in a conventional system the maximum available dynamic range is set at about 100 dB by limitations imposed by noise and the maximum permissible transmitted intensity. This dynamic range is shared between the variations in echo amplitude at particular ranges (due to the different scattering cross sections of different targets) and attenuation of echoes (which increases with distance). Compensation for attenuation can be partly provided by the application of swept gain. In practice, at any particular range, an echo amplitude variation of about 30 dB is the maximum which may usefully be accommodated, since the azimuthal resolution is unlikely to be acceptable with a larger dynamic range. Therefore, in principle around 70 dB is available to provide swept compensation for attenuation. In practice, however, swept gain cannot be sufficiently accurately applied (except by unconventional techniques: see DeClercq & Maginness 1975) over more than about 50 dB dynamic range, because of differences in attenuation along the ultrasonic beam and according to its direction.

Consideration of the frequency bandwidth of a typical pulse–echo diagnostic system indicates that 50 dB of compensation for attenuation corresponds to a penetration of about 250 wavelengths in soft tissues. Consequently, for example, the maximum penetration is around 250 mm at a frequency of 1.5 MHz, 75 mm at 5 MHz, and so on. Moreover, with a transducer diameter of 20 wavelengths (which is typical at the low megahertz frequencies), and a target dynamic range of 10 dB (at a range of 120 wavelengths), the range resolution is around 1.5 wavelengths, and the azimuthal resolution about 1.5 wavelengths (Wells 1969). The values show that it is the azimuthal resolution which limits the performance of conventional pulse–echo systems, and this limitation arises because of diffraction. The width of the ultrasonic beam in the near zone, however, can be reached over a limited range by the application of focusing. Moreover, the use of an array of transducers with appropriate time-grading allows the position of the focus to be swept along the axis of the beam, as described in §4*b*.

In addition to limited resolution, deviation from straight-line propagation (Mountford & Halliwell 1973), and high attenuation in gas and bone, one of the fundamental restrictions to the pulse–echo method in medical diagnosis is the occurrence of multiple reflexion artefacts. Multiple reflexions arise when the echo from a structure travels along an extended route by repeated reflexion between two or more interfaces. These multiple reflexions result in registrations on the display at positions where there are no corresponding reflecting structures within the patient.

(*b*) *Instrumentation*

The signal chain of a typical pulse–echo ultrasonic system is shown in figure 1. The functions of the various components have been discussed in detail by Wells (1977*a*). In summary, the clock determines the repetition rate of the system. Each clock pulse triggers the transmitter, which generates a large-amplitude voltage transient with a fast rise time. This pulse is fed through a variable attenuator (which controls the system sensitivity), to the transducer mounted in the probe. In response to this electrical pulse, the probe emits an ultrasonic pulse which travels within a beam as described in §4*a*. The backing block makes the pulse suitable for short pulse operation. Echoes returning to the probe from reflecting structures within the patient produce corresponding voltages in the transducer, and these voltage signals are amplified by the radio-frequency ultrasonic amplifier. The gain of this amplifier is electronically controlled by the output from the swept gain function generator, which is itself triggered, in this example,

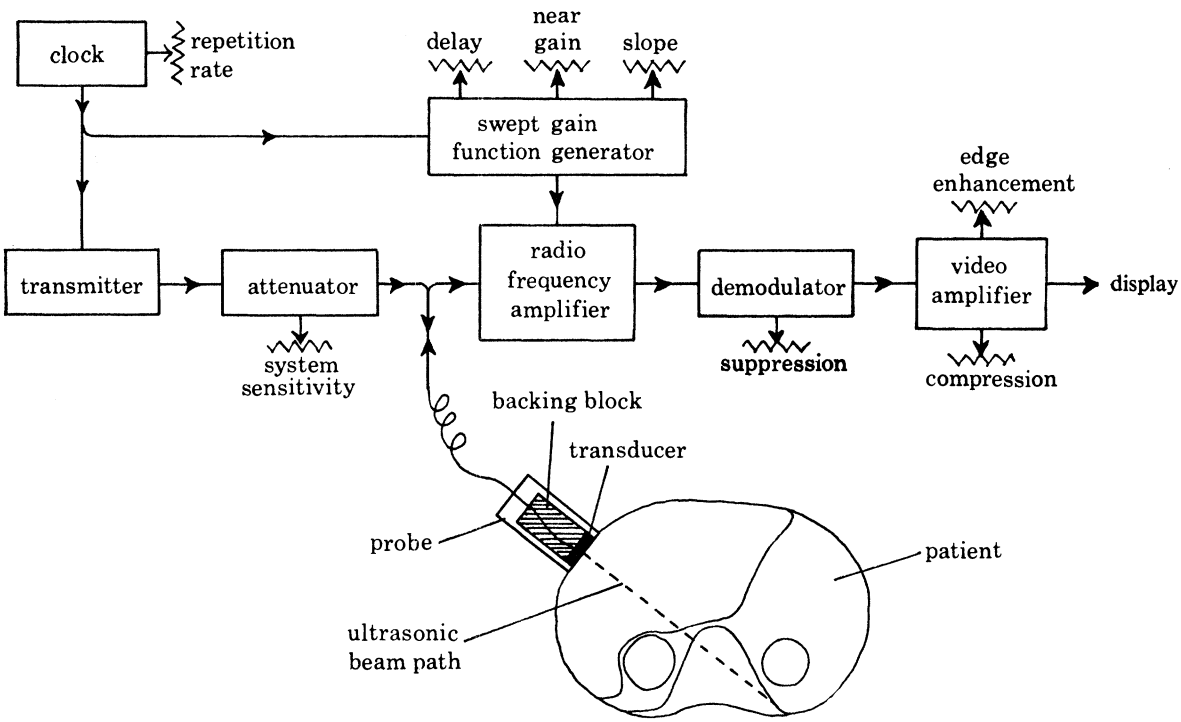

FIGURE 1. Signal chain and probe construction in a typical ultrasonic pulse–echo system.

by the clock. In this arrangement, the radio-frequency amplifier has relatively low gain at the instant that the ultrasonic pulse leaves the probe, but its gain is swept upwards progressively with time to compensate for the increasing attenuation (within the intervening tissue) of echoes originating from deeper structures. The demodulator produces a video output corresponding to the envelope of the output of the radio-frequency amplifier; small-amplitude echoes may be suppressed by reverse-biasing of the demodulator diodes. The video amplifier may have an amplitude compression characteristic (for example, a logarithmic response) to compress the useful signal dynamic at the output of the demodulator (typically 30 dB) to that which can be accommodated by the display (typically 20 dB for a brightness-modulated cathode ray tube). The video amplifier may also process the signal, for example by adding a proportion of differentiated signal to the output to enhance the leading edges of echo wavetrains.

Typical examples of various types of scanning methods and displays used in ultrasonic pulse–echo diagnosis are illustrated in figure 2. In all these displays, two separate pieces of ultrasonically derived information are presented: these are the distances along the ultrasonic beam between the probe and the echo-producing structures, and the received echo amplitudes. In the A-scope, the ultrasonic timebase is on one axis of the display, and echo amplitude is on the

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other. The probe is often hand-held, as in this example where the axial lengths of the components of the eye are being measured. (Moreover, in this example, the transducer is spaced away from the surface of the structures under examination by means of a water-filled tube: this allows echoes to be detected from the tip of the probe. The timebase and swept gain generators are triggered after appropriate delays.) There are several methods of time–position recording. In figure 2 a recording is illustrated in which the brightness-modulated timebase runs vertically downwards, while the horizontal timebase represents real time. Thus the movements of structures can be studied, in this example in the heart, by means of a hand-held probe on the chest wall.

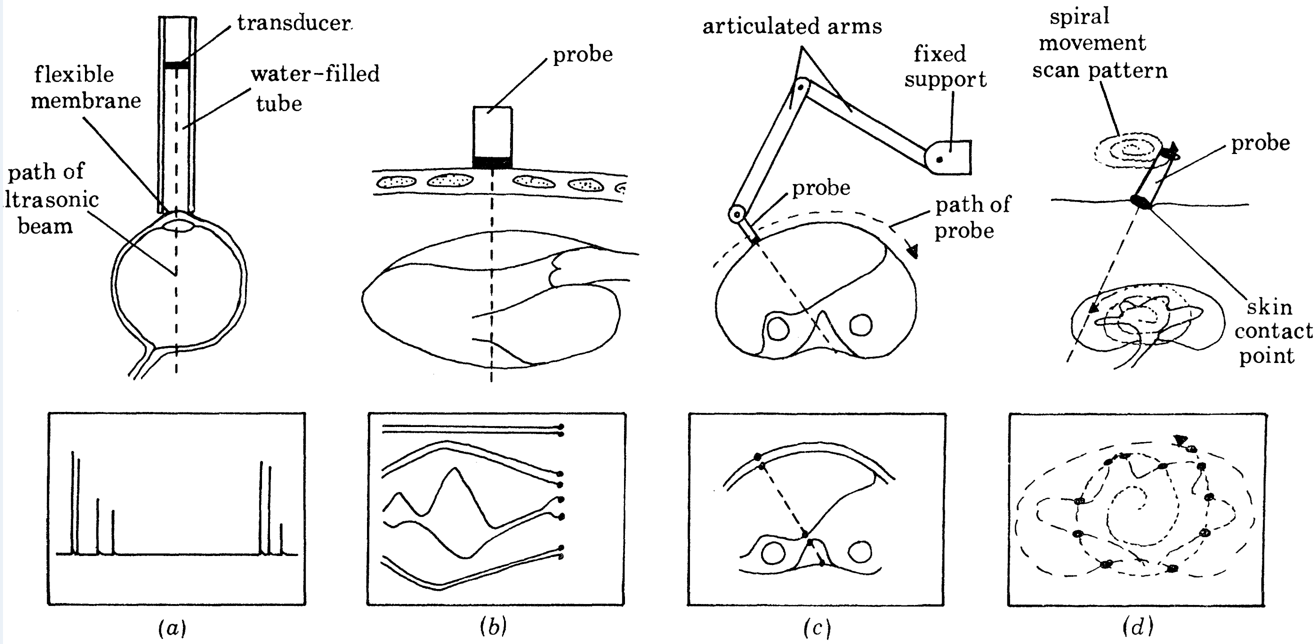

FIGURE 2. Typical scanning methods and displays used in ultrasonic pulse–echo diagnosis. (*a*) A-scan; (*b*) time–position recording (M-scan); (*c*) two-dimensional B-scan; (*d*) C-scan.

Conventional two-dimensional B-scanning, as illustrated in figure 2, depends on the movement of the ultrasonic probe across the patient. The scanning arrangement shown is one of several different designs. The horizontal and vertical deflexion circuits are driven by separate ultrasonic timebases, simultaneously triggered by the clock, as shown in figure 3. The x and y coordinates of the start of the timebases which form the image are controlled by x and y resolvers in the scanner, and the relative velocities of the two timebases are determined by a third resolver which measures the direction of the ultrasonic beam across the patient. Thus the output from the video amplifier, arranged to z-modulate the display, produces registrations on the resultant timebase in positions corresponding to echo-producing targets within the patient. A two-dimensional image is built up by moving the probe so that the part of the anatomical cross section which it is desired to visualize is scanned by a sufficient number of discrete ultrasonic lines.

The formation of a clinically useful two-dimensional B-scan requires either that it is completed in a short interval of time or that it is stored. The simplest method of storage is by photographic film, recording the image with an open shutter while it is displayed on a conventional cathode-ray tube. This method has two principal disadvantages: these are the delay and expense of photography, and the danger of 'overwriting' leading to non-uniform pictures when the scanning motion is irregular. Direct-view electronic storage tubes have very low dynamic range (the bistable type has zero dynamic range), and so they are not suitable for high-quality imaging which requires a grey scale. The use of a scan converter as the storage system goes a long way towards overcoming these difficulties. The analogue type of scan converter (Deschamps & Kleehammer 1977; figure 3) resembles an electromagnetically deflected cathode-

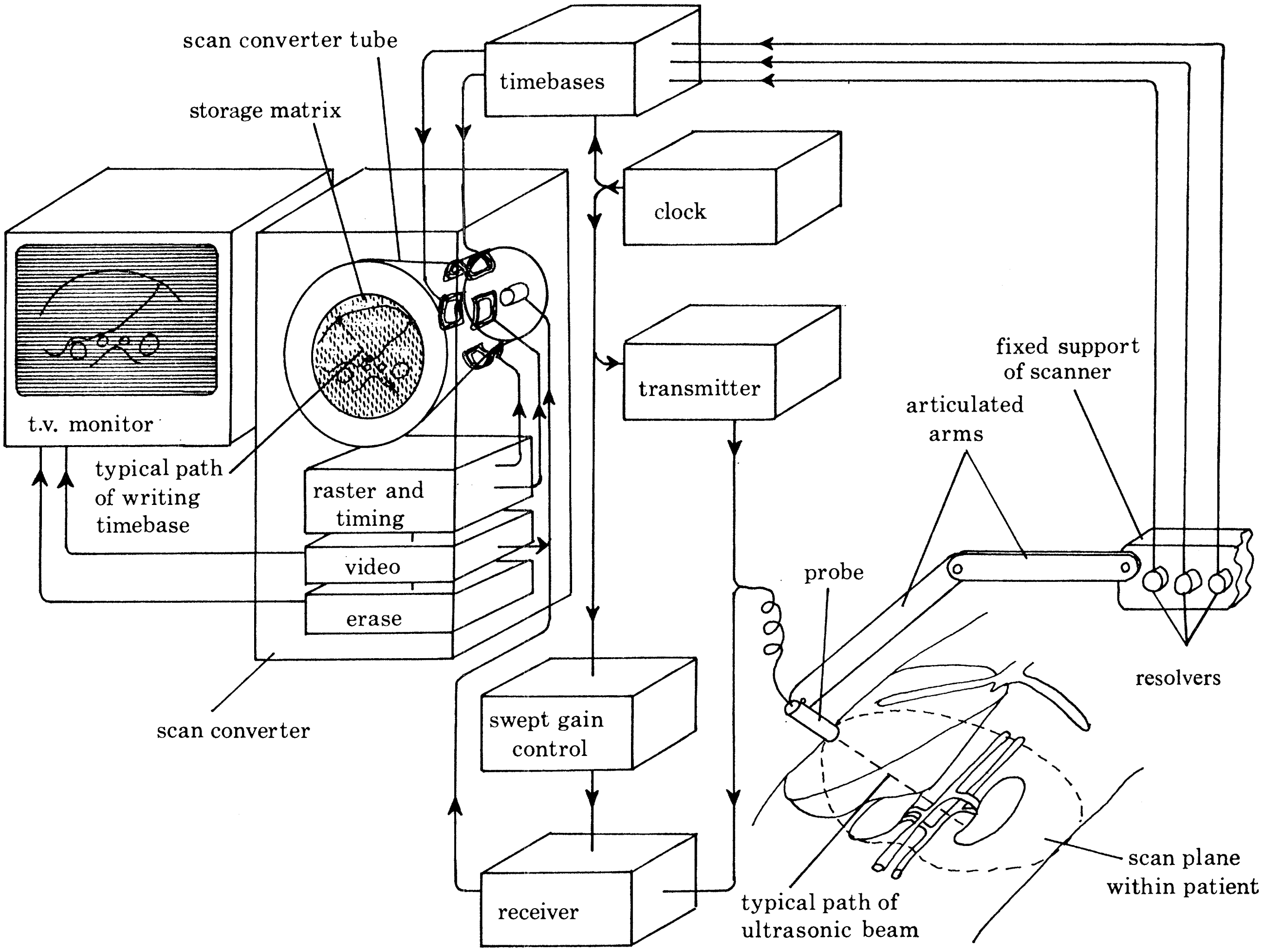

FIGURE 3. Typical manually operated conventional ultrasonic two-dimensional B-scanner.

ray tube, except that the phosphorescent screen is replaced by a small target consisting of a silicon backplate covered with a mosaic of tiny silicon oxide storage elements. During scanning, the scan converter is operated in the equilibrium writing mode, and a charge pattern representing the image is built up, in which the charge on each pixel is related to the maximum echo amplitude received from the corresponding point within the patient. In the reading mode, the storage matrix is scanned in a t.v. raster to produce an image on an ordinary cathode-ray tube monitor. By suitable time-sharing, writing and reading can be interlaced so that the picture can be observed while the patient is being scanned. This type of scan converter has a resolution of better than 1000 lines, a dynamic range of more than 20 dB, and a readout viewing time of more than 10 min. Were it not for the fact that drift occurs so that optimum settings are difficult to maintain over a long period, and individual lines cannot be selected for updating, the analogue scan converter would provide an ideal method of storing ultrasonic images. It is in order to overcome these limitations that digital scan converters (Ophir & Goldstein 1977) have recently been developed. A typical modern digital scan converter has 512 × 512 5-bit words, operated as a random access memory. The principal disadvantage of this arrangement is that the pixel size, although small in relation to the ultrasonic resolution, nevertheless introduces tiresome quantization steps in the positional registration of the image, and interpolation is necessary to fill in occasional empty pixels.

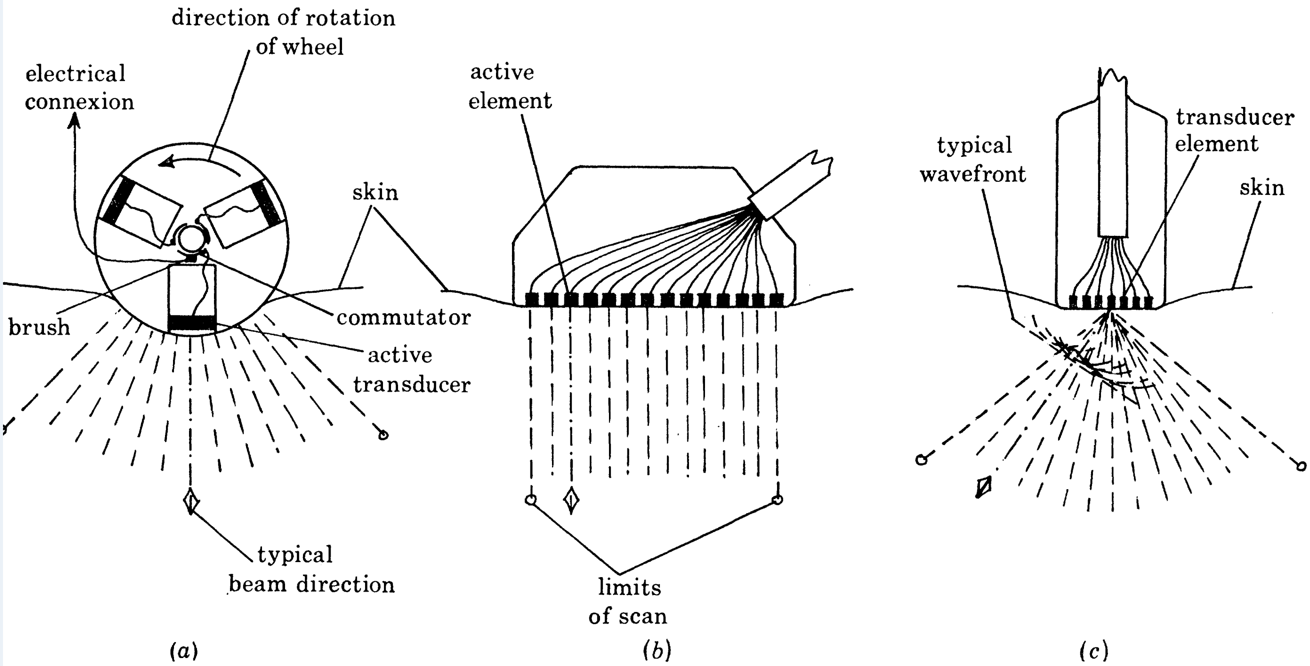

FIGURE 4. Typical rapid-scanning methods for two-dimensional imaging in real time. (*a*) Mechanical system: continuously rotating wheel with commutated transducers, producing a sector scan. (*b*) Linear array: elements operated singly or in groups, producing a rectangular scan. (*c*) Steered array: controlled timing gradient across the array producing a sector scan.

The remaining method of investigation illustrated in figure 2 is C-scanning. The probe is moved in a regular pattern (for example, in a conical path with increasing base radius, or in a raster), and a plane at a fixed distance from the plane in which the transducer lies is interrogated by appropriate time-gating. A wide aperture can be used, with strong focusing to obtain high resolution and to minimize the effect of specular reflectors (Mezrich 1977). Although the method is not at present used in routine clinical practice, it makes it possible in principle to visualize otherwise inaccessible planes (such as the lateral section of heart), and the technique deserves more attention.

An important class of B-scanners uses water-bath coupling between the transducer and the patient. The best of the modern machines (Kossoff *et al.* 1976) has eight large aperture transducers, used in sequence automatically to produce excellent images in less than about 3 s per frame.

The rate at which ultrasonic pulse–echo data can be obtained is limited by the time required to penetrate to the necessary depth within the patient. For example, for a penetration of 150 mm, the echo acquisition time is about 200 μs per line. Consequently, pulses can be transmitted at intervals of 500 μs (or a little less) without introducing artefacts: the corresponding pulse repetition rate is 2000 s^{-1}. The pulse repetition rate is equal to the product of the number of lines per frame and the image frame rate; thus, if an image frame rate of 50 s^{-1} is required, each frame is composed of 40 lines, and so on. This high rate of data acquisition makes possible the operation of yet another class of two-dimensional B-scanners which have high image rates, called (sometimes inaccurately) 'real-time' systems. The principles of some of these types of instrument are illustrated in figure 4. High-speed mechanical scanners (see, for example, Holm *et al.* 1975) are simple in principle, but they lack versatility and some are difficult to couple to the patient. Multi-element systems, although requiring more complicated electronics and generally having poorer resolution, sidelobe performance and near-field capability, are more versatile and easier to couple. There are two distinct types of array system. The first employs a linear array (Bom, Lancée, Honkoop & Hugenholtz 1971) in which single elements, or groups of elements, are used in sequence rapidly to step the ultrasonic beam along the array. The second type (Somer 1968) employs an array of very narrow strips, and appropriate time delays are introduced across the array to steer the beam in any desired angle within a sector. (This is usually but inappropriately called a 'phased array'. Alas, it is now probably too late to change the terminology; strictly, a phased array, as used in radar, operates over a much narrower frequency band relative to the centre frequency than is the case in ultrasonic diagnostics.)

In ultrasonic diagnosis, real-time imaging has three main advantages in relation to conventional two-dimensional scanning. Thus, rapid physiological events can be studied, physiological movements do not degrade the image, and the operator can more quickly interpret anatomical relations. The term 'real-time' applies to any imaging system which produces images rapidly enough to allow changes in the scanned plane to be presented to the observer without a time delay significantly affecting the perception of the changes. If this requires, in a particular situation, an image rate of less than about 16 s^{-1}, flicker can be avoided by the use of a scan converter with line-by-line refresh capability.

(*c*) *Clinical applications*

Ultrasonic diagnosis is already in routine use in almost every major clinical speciality, and in some minor ones as well. The most important applications and recent developments are mentioned briefly here with some representative references, but detailed descriptions of work up to the end of 1975 can be found in Wells (1977*a*, *b*).

In *angiology*, the principal application is in the visualization of the abdominal vessels, especially the aorta and aortic aneurysms. The resolution of modern scanners is good enough to display even the mesenteric and renal arteries (Leopold 1975). Peripheral vasculature is well shown by high-resolution real-time scanning (Green *et al.* 1977). The major applications in *cardiology* are in the study of structure movement by time–position recording, and in two-dimensional real-time scanning (Gramiak & Waag 1975; Bom 1977; Roelandt 1977). In *endocrinology*, the structure of the thyroid gland is readily displayed (Kobayashi 1977). There are many applications in *internal medicine*, ranging from structural studies of anatomical relations, through diagnosis of pancreatic disease, to incorporation within diagnostic algorithms

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(Goldberg 1977). The early promise in *neurology* has sadly proved often to be false, but there are some interesting applications in children, in whom the skull does not intolerably distort the ultrasonic field (Garrett, Kossoff & Jones 1975). At present, probably the largest proportion of ultrasonic examinations is in *obstetrics*, where the method is invaluable in determining maturity, detecting multiple pregnancy and locating the placenta. Recent progress has included studies of foetal breathing movements (Boddy & Dawes 1975). Many opportunities for widespread introduction of ultrasonic techniques now exist in *oncology*, ranging from tumour localization and staging to radiotherapy treatment planning (Jentzsch, Kärcher & Böhm 1974). Ultrasonic techniques have been used for many years in *ophthalmology*, having their principal applications in biometry (Giglio, Ludlam & Wittenberg 1968) and in two-dimensional studies of the eye and orbit (Sutherland & Forrester 1974). Finally, among the major specialities, in *urology* ultrasonics is useful in the investigation of space-occupying renal lesions, and in the diagnosis and staging of bladder and prostate tumours (Barnett & Morley 1972; Whittingham & Bishop 1973).

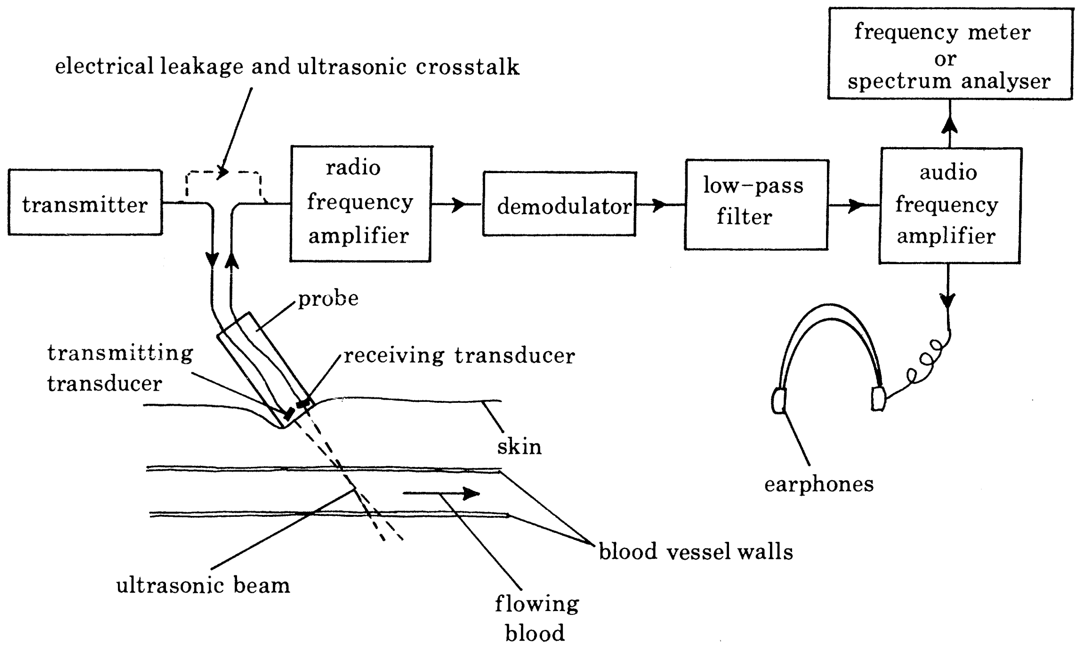

FIGURE 5. Block diagram of simple continuous-wave ultrasonic Doppler blood flow detector.

5. THE DOPPLER METHOD

Almost all medical applications of the ultrasonic Doppler effect are based on reflexion (Woodcock 1975). The Doppler shift f_D in the frequency of reflected ultrasound may be used as a measure of the velocity vector of the movement of the reflecting surface or ensemble. If γ is the angle between the direction of flow and the effective axis of the ultrasonic beam,

$$v = -f_D\, c/(2f\cos\gamma), \tag{11}$$

where v is the flow velocity and $c \gg v$. For example, $f_D = 1$ kHz when the measurement is made in blood flowing at 100 mm s^{-1}, and $f = 8$ MHz and $\gamma = 15°$.

A simple continuous wave Doppler system is illustrated in figure 5. The transmitter is a power oscillator which drives the transmitting transducer at constant frequency. Echoes returning from within the patient are detected by the receiver transducer, mounted beside the transmitting transducer. They have frequencies equal to that of the transmitter if they originate from stationary reflectors, but shifted by the Doppler effect with reflectors with a component of velocity along the ultrasonic beam.

According to the application, the operator may simply listen to the output from the Doppler detector, or it may be processed to give numerical data, for example by zero-crossing frequency meter or by frequency spectrum analysis. More complicated Doppler instruments have directionally sensitive detectors, and so can distinguish between forward and reverse flow.

Having been established for many years in unsophisticated clinical applications such as the detection of the foetal heart (Liu, Thomas & Blackwell 1975), and simple investigation of peripheral vascular disease (Fitzgerald, Gosling & Woodcock 1971), the full potential of the Doppler method is only now beginning to be appreciated. Continuous wave Doppler instruments are being used to produce two-dimensional scans of blood vessels (White & Curry 1976). Pulsing the system allows Doppler and range information to be gathered simultaneously (within the limitation imposed on range resolution by the conflicting requirements of bandwidth: see Atkinson & Wells (1977), thus allowing transverse sections through flowing blood to be obtained (Fish, Kakkar, Corrigan & Nicholaides 1972).

6. Newer methods

Ultrasonic investigation of the brain through the intact adult skull previously posed apparently intractable problems, but recent results with phase-compensation of arrays are quite promising (Smith, Phillips, von Ramm & Thurstone 1976).

The possibility of extending the horizons of conventional pulse–echo ultrasonic imaging beyond anatomical mapping and grey-scale presentation to the quantitative analysis of echo signals is being actively investigated (Linzer 1976).

Although the early attempts to visualize anatomical structures by transmitted ultrasound failed, recently time delay spectrometry (Heyser & le Croissette 1974) and incoherent ultrasound (Havlice, Green, Taenzer & Mullen 1977) have given encouraging results in shadow imaging. Another method which is being studied is computerized tomography by reconstruction from profiles of transmitted amplitude or time of flight (see, for example, Glover 1977). Serious problems remain to be solved in attenuation tomography, but velocity tomography of the breast has given interpretable images, and indicated the generalized nature of breast cancer.

In the field of Doppler ultrasound, more powerful methods are being applied to transfer function analysis in the assessment of peripheral arterial disease (Skidmore & Woodcock 1978). Another interesting development is the discovery of the distinctive signals obtained from blood flowing in the neovascular channels associated with malignant tumour growth (Wells *et al.* 1977).

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Discussion

H. A. B. Simons (*Royal Free Hospital School of Medicine, Pond Street, London NW*3 2*QG, U.K.*). Why has there been such a long delay in the introduction of ultrasonics into medicine when the fundamental physics is well known? Very considerable technical work in the transmission of sound in water was done during the last war, as, for example, in submarine detection and acoustic mines.

P. N. T. Wells. I think that there are two principal reasons for this. The first is that the senior professional men at that time made very negative statements about the promise and feasibility of pulse–echo ultrasound methods. The second is that it really is only recently that the thermionic valve has been replaced by the semiconductor device.

J. Mallard (*Department of Bio-medical Physics and Bio-engineering, University of Aberdeen, Aberdeen, U.K.*). Following on from one of Dr Shepstone's points, we have been doing computer reconstructed tomographic imaging with radionuclides on patients since 1971, some time before the E.M.I. scanner using X-rays was announced. We have used it to produce transverse section scans from isotopes in many different organs, including the brain. For our clinical series of brain scans, we have found that our overall diagnostic accuracy has risen from 80 % without the section image to 96 % with the section, in other words, our errors have dropped from 20 % to 4 %. For the detection of brain tumours alone our accuracy has risen from 80 % to 90 %; that is, our errors have fallen from 20 % to 10 %. This is an extremely significant improvement in nuclear medicine diagnostic accuracy and the scanner, which can produce conventional views, as well as the section views, is now available commercially.

Following on from Professor Simons's point just now, and Dr Wells's and Professor Gordon's point earlier, I think there are two other factors contributing to the comparatively slow adoption of ultrasound in medicine. The very rapid adoption of X-rays was before we knew anything about the harmful effect of X-rays, and it was only after 50 years that the damaging effects were understood. We can expect the medical profession to be suspicious of a new form of radiation, and even though no one has conclusively demonstrated any deleterious effects at diagnostic levels of ultrasound radiation, the suspicion is still there.

Also, human beings have developed over millions of years, I believe, to cope efficiently and rapidly with images, and the eye and brain are magnificently adapted for this task. It is only comparatively recently that ultrasound has been used to produce pictures, rather than the less-easily understood graphs or blips on oscilloscopes. Only in the last 10 years have we had the B-scan images, which are now rapidly being taken up; only in the last year have we had the images of flow which Dr Wells showed in his own carotid arteries.