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The mechanical behaviour of human tissue

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The human body comprises several different forms of tissue. Understanding both the normal and abnormal mechanical behaviour of different tissues can be of considerable help to both the physician and the surgeon. To achieve this understanding requires the closest of collaboration amongst doctors, engineers and materials scientists in the form of joint research projects. This paper indicates some of these topics in the hope of illustrating the medical value, from the patient's point of view, of the results of this form of multidisciplinary research. This field of activity is very broad and no single review can hope to be comprehensive. However, the author hopes that the topics described will encourage other interested materials scientists to pursue research in this subject. More detailed information can be found in the individual references.

1. Introduction

All material on the surface of the earth is subject to the effects of gravity and is in a state of stress. The human body is no exception to this rule and indeed, with the coming of the space age, the body is exposed in many cases to exaggerated inertia forces and corresponding stresses. The body is also a dynamic system and, as such, has to respond to continually changing force actions. A simple illustration of the magnitude of such force actions obtains in the normal state of walking, where at "heel strike" the effective force on the hip joint can be of the order of six to seven times the body weight, owing to dynamic effects.

Over the years, many investigators, both medical and non-medical, have carried out experimental and theoretical analyses of the body and the mechanical behaviour of its tissues. As far back as 1680, Borelli [1] was measuring the strength of jaw muscles by hanging weights on the lower jaw. Two hundred years earlier, Leonardo da Vinci was, by dissection, describing in considerable detail the anatomical structure of the body.

To engineers and materials scientists, the analysis of stress in a material must usually be preceded by the analysis of the forces producing that stress. The same is true of the human body and much of the earlier work was devoted to the analysis of force actions on the body rather than studying the mechanical strength of its tissues. Perhaps one of the earliest recorded studies of the mechanical behaviour of tissue was the work done on the behaviour of skin carried out by Langer (1861) [2]. His experiments involved making a large number of stab wounds with a round-bladed dagger in corpses. The resulting wounds took up an elliptical shape and the major axes of these ellipses formed a pattern of lines which were regarded as lines of tension. Unfortunately, many of these lines run transverse to natural creases in the skin and cannot, therefore, be regarded simply as tension lines. Even today there is still no satisfactory explanation of this phenomenon. More recent work on the mechanical behaviour of skin which is discussed later does, however, provide a better understanding of some of the factors involved in the response of skin to stress.

The present paper is aimed at providing an introductory background for those interested in the mechanical behaviour of materials both from the micro- and macroscopic point of view. For this reason, and since human tissue takes many different forms, I do not intend to examine in detail one particular tissue. Instead, a description will be given of work which has been carried out on a number of different tissues These are: (i) skin, (ii) costal cartilage, (iii) oesophageal tissue, (iv) cardiovascular tissue, and (v) bone.

In most cases, I hope to show that the value of the research is not merely academic but rather that the results can be applied usefully in medical practice. If it is accepted that the study of human tissue, whether by materials scientists or engineers, is in the field of biomedical engineering, then the aim of that field of study should be noted: biomedical engineering is the application of the methodology of the physical sciences and engineering to medicine in relation to the care and treatment of the human patient.

2. The mechanical behaviour of skin

As already noted in the introduction, the first recorded serious studies of skin behaviour under stress were made in 1861. These studies were not added to for almost a hundred years as it was only in the late 1950s that more precise investigations into the mechanical behaviour of skin were undertaken. During the second world war, plastic surgery developed rapidly to deal with the many cases of disfigurement arising from war injuries, particularly in relation to facial burns. After the war, this form of surgery became more routine and surgical intervention to remove disfigurement for cosmetic reasons became more popular.

Plastic surgeons soon realized that their procedures might be improved if they could predict more accurately the way in which skin responded to applied stress. To obtain this information they sought the help of scientists who were familiar with the techniques required for the study of the mechanical behaviour of materials, initially engineers and later materials scientists. Detailed investigations by Kenedi *et al* [3] were carried out on detached specimens of skin from subjects of different ages and taken from various parts of the body.

Typical stress/strain relationships for such specimens taken from the abdominal region are shown in Fig. 1 in comparison with the curve obtained from tensile tests on isolated collagen fibre. This latter material, which is the principal constituent of skin, provides its tensile strength and is defined as a fibrous protein. Fig. 1 illustrates clearly important phenomena in the mechanical behaviour of skin. These are:

(i) initially large strains are produced for small load;

(ii) at some stage the complete opposite of this behaviour arises and small additional strains are produced by large increases in load;

(iii) the stage at which the change from (i) to (ii) occurs varies progressively with age, extensibility being greatest in the newborn child, and

(iv) the modulus in stage-(ii) behaviour tends to the value obtaining for isolated collagen fibre.

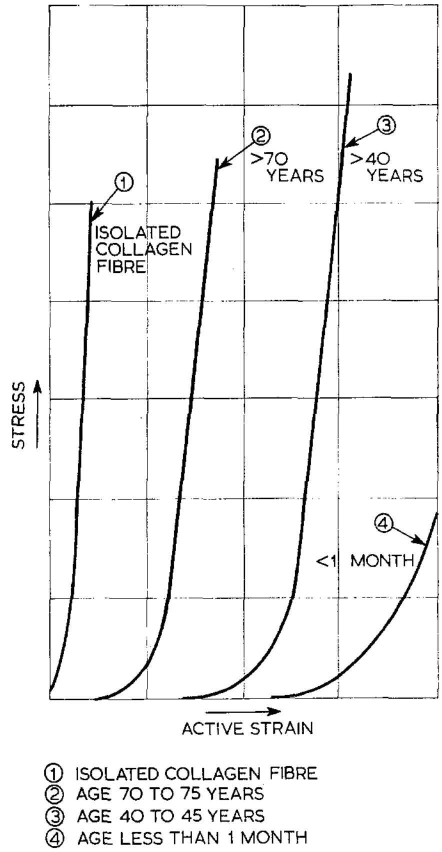

Figure 1 Stress/strain relationships for detached specimens of skin and isolated collagen fibre.

Perhaps the most significant result from these investigations is the phenomenon indicated in (iv) above. This experimental result has been explained by histological studies of unstressed and stressed skin. These studies have shown that in the unstressed stage the collagen fibres are randomly orientated, whilst in the stressed condition they are aligned in the direction of the active strain, that is, in the direction of the applied stress. The same histological studies, carried out by Craik and McNeil [4], also showed that after the application of low stress levels the collagen fibres returned to their random orientation on removal of the stress.

The brief description given so far of tensile testing of skin has not dealt with the more

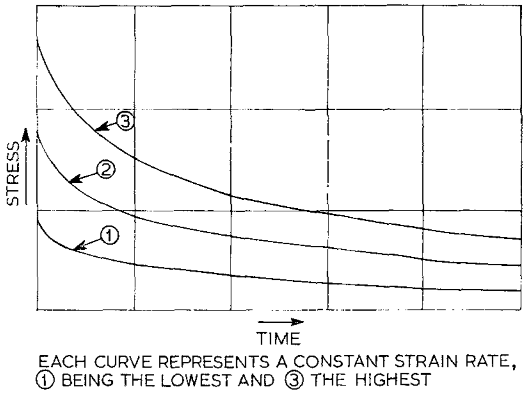

Figure 2 Stress relaxation behaviour of detached skin specimens.

analytical investigations which have been undertaken by researchers such as Crisp [5] and Fung [6]. Skin, in common with other biological material, behaves viscoelastically, as shown by the stress/time curves at different stain levels for tensile tests on detached skin illustrated in Fig. 2.

In this present paper it is not intended to deal with the more complex aspects of tissue behaviour. However, the stress relaxation capacity of skin arising from its viscoelastic behaviour is not merely of academic interest. For example, the plastic surgeon, equipped with adequate knowledge of how much extra tension can be applied to skin, is able to adapt his operating procedures on the basis of that knowledge.

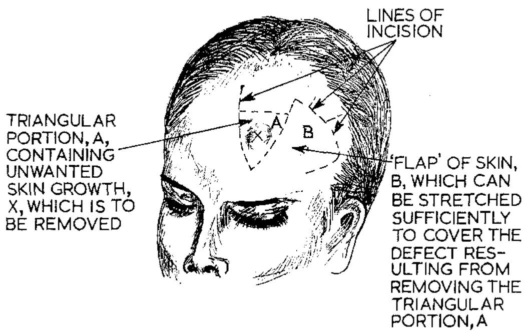

Figure 3 The removal of a skin growth from the forehead; pre-operative state.

The illustration of this process is shown in Figs. 3 and 4. In this case the plastic surgeon is required to remove an unwanted and unsightly skin growth from the forehead. This involves removing a finite area of skin as shown in Fig. 3 (portion A). To repair the resulting defect might have required the grafting of a piece of skin taken from some other part of the patient's body. This latter procedure can often yield a result which is cosmetically unacceptable. However, by knowing how much extra tension can be taken by forehead skin, the "stretched flap" technique, indicated in Fig. 3, can be used. The resulting appearance after completing the procedure is shown in Fig. 4. In time the suture lines heal and the final result of this surgical intervention is cosmetically acceptable.

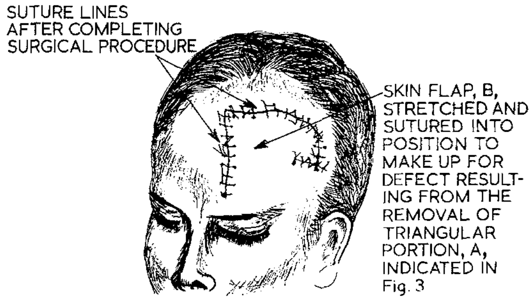

Figure 4 The removal of a skin growth from the forehead; post-operative state.

3. Residual stresses in costal cartilage

In the previous section on skin, attention was drawn to the viscoelastic behaviour of biological tissue. Another example of such behaviour has been found from compression tests on costal cartilage carried out by Abrahams and Duggan [7]. The typical form of the stress/time relationships which they obtained for the compression of costal cartilage is shown in Fig. 5. The vertical axis represents a dimensionless stress parameter defined by the ration of stress, σ_t, at time t, to the initial stress, σ_0.

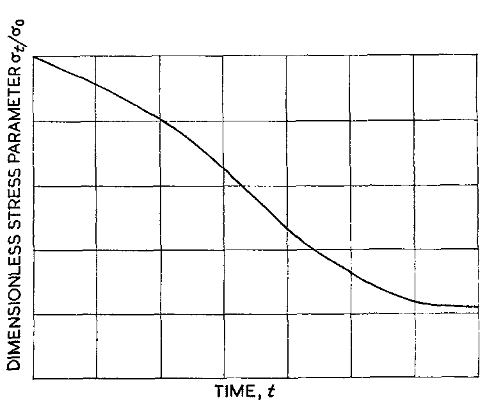

Figure 5 Stress relaxation behaviour of costal cartilage.

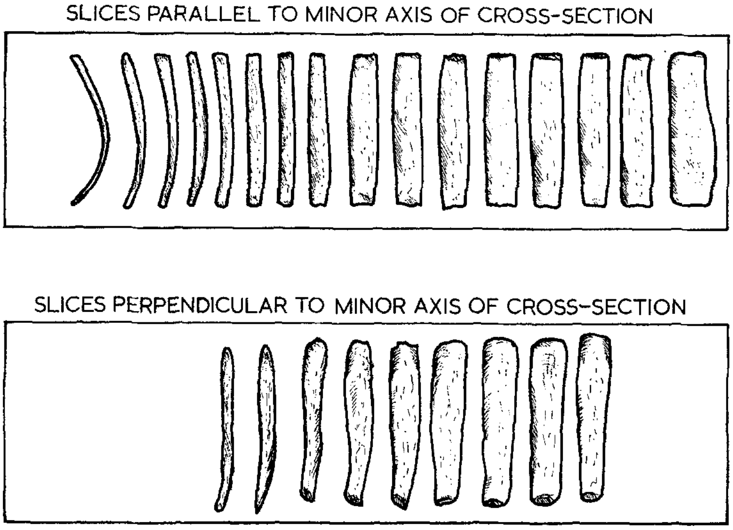

Figure 6 Deformation behaviour of slices of costal cartilage cut parallel and perpendicular to the minor axis of the specimen cross-section.

Costal cartilage is the particular cartilage which connects some of the ribs to the sternum (breastbone). In general, cartilage comprises a resilient matrix containing polysaccharide, together with a large number of collagen fibres. As a homograft material it is often used for the repair of nasal bridge and other skeletal defects. It can be carved to any desired shape and is particularly convenient for insertion in the nasal bridge. Unfortunately, early experience of its use in the repair of nasal bridge defects sometimes gave rise to an unwanted deformation of the graft after implants. This deformation resulted in nasal distortion which became evident immediately the splints used to fix the nose in position after surgery were removed.

An explanation of this phenomenon has been given by Kenedi *et al* [8] which employs the concept of a balanced system of self-locked stresses (internal stresses in materials science terminology) which exist in the cartilage when in its intact costal position. Removal from its natural position and carving to appropriate shape disturbs the stress balance in the cartilage. This produces distortion owing to the redistribution of stresses required to attain a new balanced state.

The analysis of this phenomenon, carried out by Abrahams and Duggan, involved measurements made on samples of costal cartilage sliced parallel and perpendicular to the minor geometric axis of the roughly elliptical cartilage specimen cross-section. Typical results of this procedure are shown in Fig. 6. The shapes on the left correspond to slicing parallel to the minor axis and those on the right to slicing perpendicular to the minor axis. The increasing curvatures developed by the parallel slices indicate the re-distribution of the self-locked stresses.

The tests enabled the researchers to measure curvatures and corresponding cross-sectional dimensions at different stages of slicing. These measurements could then be used to build up a picture of the self-locked stresses originally obtaining in the intact specimen if an effective mean value of Young's modulus could be found for the material. Such a value was determined experimentally from tests involving increasing compression or tension where stress values were recorded after complete relaxation of each strain increment. A typical self-locked stress distribution for a costal cartilage specimen cross-section determined by this method is shown in Fig. 7.

This form of stress distribution was found to be reasonably repeatable over a range of specimens. As a result, a technique was developed by Abrahams and Duggan which enabled the plastic surgeon to carve the cartilage to the required depth and still have a balanced self-locked stress system. The resulting homograft exhibited no appreciable distortion after implant.

This particular example of the mechanical behaviour of human tissue illustrates a comment

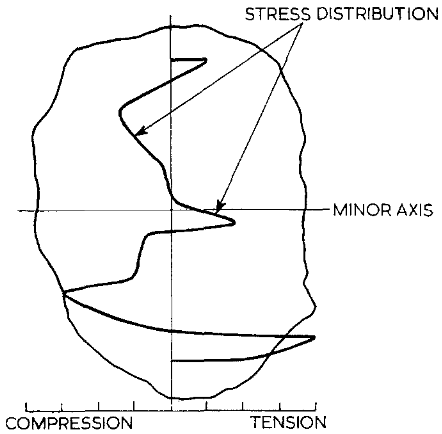

Figure 7 Estimated distribution of self-locked stresses in a costal cartilage specimen cross-section.

made recently by Fung [6] that "a history-dependent element exists in the mechanical action of all biological materials" and that "the stress at a given point at any instant of time depends not only on the strain at that time, but also on the strain history".

4. The anisotropic behaviour of oesophageal tissue

The structural form of some human tissue can often give rise to a particular form of material behaviour. An example of this is the anisotropic behaviour of eosophageal tissue.

The oesophagus is a hollow muscular organ lined by epithelium. The epithelium is for the most part squamous*, but near the stomach there is a sharp demarcation at the so-called Z-line, to columnar* epithelium. There are only occasional mucus glands and, as the role of the oesophagus is the transport of food and liquids from the mouth to the stomach, the epithelium acts as a protective inner lining.

The working part of the oesophagus is the muscle in its wall, which propels the food into the stomach by a peristaltic action. At the upper end the muscle is all striated or voluntary muscle, whilst at the lower end it is all smooth or involuntary muscle; both are present in the middle. There are two muscle layers, an inner helical and an outer longitudinal, but muscle fibres interweave so as to form a quilt-like pattern. It is this twin muscle layer system which produces the anisotropic behaviour.

Tensile tests on post-mortem specimens of oesophageal tissue have been carried out by Black and Melcher [9] as part of a research project on the development of an artificial oesophageal replacement. Such a prosthesis

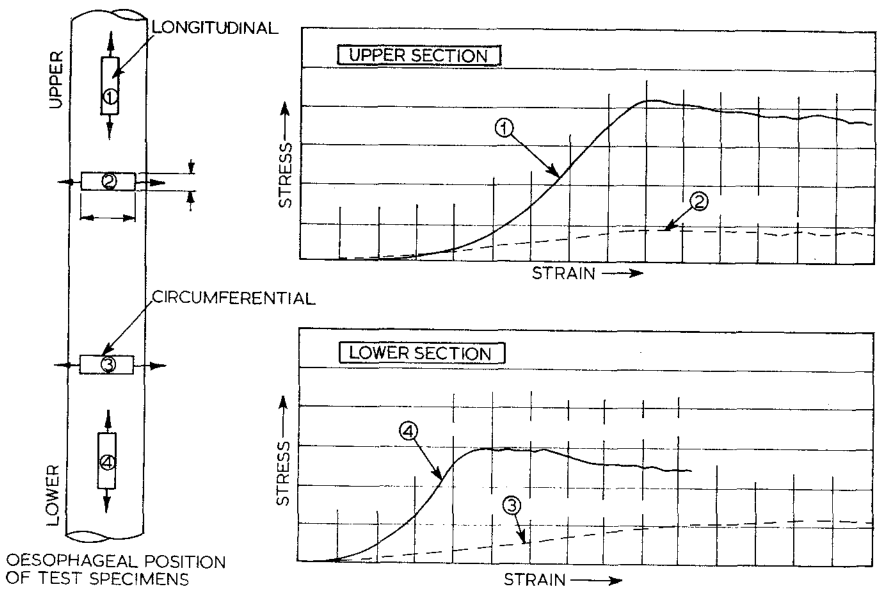

Figure 8 Stress/strain relationships for longitudinal and circumferential specimens of post-mortem oesophageal tissue.

*Squamus and columnar are terms used to describe the cell structure of the epithelium.

would be of value in cases where the lumen of the oesophagus is occluded owing to a benign or malignant growth.

The tests were performed at constant strain-rates with the specimens immersed in a saline solution. Specimens were taken from the upper and lower ends of the oesophagus in both the longitudinal and circumferential directions. Typical results, in the form of stress/strain curves obtained from these tests, are shown in Fig. 8. As expected, the significant feature of these test results is the much larger extension per unit stress which obtains for specimens cut in the circumferential direction. This phenomenon occurs irrespective of the position in the oesophagus from which the samples are taken and demonstrates clearly the anisotropic behaviour of this particular tissue. It is of interest to note that, unlike skin behaviour, the results of approximately one hundred oesophageal test specimens indicated no correlation whatsoever between material behaviour and age.

The results of these tests might suggest that any artificial replacement tube should have similar anisotropic properties. This could be achieved by appropriate fibre-reinforcement placement.

However, it must be remembered that the biological oesophagus transports ingested material by a muscular action which cannot be reproduced by an artificial tube. The passage of ingested material through the replacement tube will have to rely on the force of gravity.

Thus, initially, the most important requirement for the artificial tube is that it has a wall stiffness sufficient to prevent collapse of the tube under the differential pressure effects to which it is subjected within the body. This wall stiffness requirement is likely to militate against the development of an anisotropic tube wall even although its biological counterpart displays such pronounced anisotropic behaviour.

Other parts of the body have also been found to behave anisotropically to a significant degree. For example, the ratio of longitudinal to circumferential tensile strengths of some parts of the aorta, the main artery of the body, has been quoted by Atsumi and Sakurai [10] to be in the region of 4:1. The mechanical testing of arterial and other tissue components of the cardiovascular system is of considerable interest to the cardiologist and thoracic surgeon and is discussed in the section which follows.

5. Cardiovascular tissue

The haemodynamic behaviour of the cardiovascular system has been shown by many researchers to be highly dependent on the elasticity of the arterial walls. This subject has received considerable attention from several mathematicians and their work has been summarized clearly by Rashevsky [11]. This latter work has been related to the possibility of determining the elasticity of blood vessels in intact living humans.

The potential use for this kind of information is related to clinical studies of arteriosclerosis and associated diseases. Arteriosclerosis is a widespread disease of ageing which is characterized by infiltration of cholesterol into certain lesions of the arterial walls, distorting the vessels and making them rigid. The condition itself is serious and also produces a predisposition in the patient to heart attacks, cerebral thrombosis and other serious illnesses.

From the materials scientist's point of view, arteriosclerosis involves a change in the mechanical properties of the blood vessel walls. This change can either be described in terms of the "volume elasticity" of the blood vessel or the value of "Young's modulus" for the vessel wall. The "volume elasticity" is defined as the inverse of the factor of proportionality between the pressure in the vessel and the increase in volume produced by the pressure. This means that the value of the "volume elasticity" increases as the artery becomes less elastic. The limitation of this definition is that it assumes a linear pressure/volume relationship for the blood vessel. However, experimental determinations of the volume elasticity of excised pieces of aorta carried out by Remington *et al* [12] indicated a non-linear pressure/volume relationship. It must be noted that this form of *in-vitro* testing introduces the additional problem of the difference in mechanical behaviour between dead and living tissue. This particular difficulty arises frequently in the study of the mechanical behaviour of tissue and any extrapolation of *in-vitro* test results to the *in-vivo* situation can only be made after careful assessment of the histological changes in the tissue which occur after removal from the body.

It has been shown by Rashevsky that the "volume elasticity" is proportional to Young's modulus for the blood vessel wall. However, it also depends on the initial thickness of the vessel wall and the initial volume. Thus "volume elasticity", unlike Young's modulus, is not a material constant.

The above comments are only a brief indication of the importance of understanding the mechanical behaviour of blood vessel walls in explaining both normal and abnormal phenomena in the performance of the human cardiovascular system. Amongst the more detailed treatments of this subject, those presented by McDonald [13] and Anliker [14] indicate the more significant aspects of this area of research.

The mechanical testing of all biological tissue presents many problems, not least of which is the preparation and gripping of thin tissue specimens. In relation to some cardiovascular tissue such as heart valve leaflets, this problem is even more pronounced. In recent years the use of homo- and heterograft valves (that is, valves taken from post-mortem humans or animals, respectively) and fresh autologous fascia lata valves* to replace severely damaged and incompetent human heart valves has been attempted. The success rate of such procedures is for many heart surgeons still not satisfactory as there are still too many failures both early and late, due to mechanical failure of the replacement valve tissue. The methods used to sterilize and preserve such valves prior to implant are thought to have a considerable effect on the ability of the valve leaflets to withstand repeated mechanical stress when finally placed in the recipient heart.

In order to study these problems and obtain quantitative information for comparative purposes it is necessary to determine stress/strain relationships for the valve leaflet tissue. The extreme thinness of this tissue makes it difficult to perform stress/strain tests by normal procedures. An interesting stress/strain testing technique which overcomes this problem of leaflet thinness has been described by Mundth *et al* [15]. Basically, their test procedure involved pressurizing a tissue diaphragm by means of a constant volume infusion pump. Continuous recording of the pressure using a suitable transducer enabled the authors to record pressure/volume curves for the tissue diaphragm specimens. Analysis of these results produced values of meridional stress and corresponding meridional strain at different stress increments. This approach enables the calculation of the "incremental modulus of elasticity" which represents the value of Young's modulus corresponding to particular increments of stress. Values of this parameter can be used for a quantitative comparison of different leaflet and blood vessel tissues. A diagrammatic representation of the test equipment is shown in Fig. 9 which illustrates the relative simplicity of this particular form of materials testing.

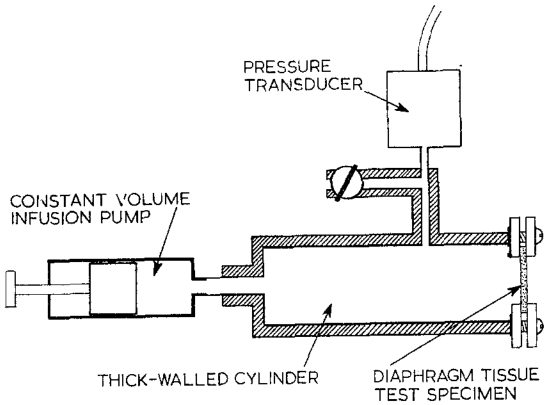

Figure 9 Test equipment for determining stress/strain relationships for cardiovascular tissue using circular diaphragm specimens.

It is of interest to note that the results obtained by Mundth *et al* for valve leaflet tissue were similar to those already described for skin. Thus, initially, there was a large extension for low stress followed by a region of small extension for high stress.

The results of these and other investigations into the behaviour of cardiovascular tissue have helped both cardiologists and thoracic surgeons in developing their diagnostic techniques. They have also been of great importance to the design and development of mechanical simulators of the heart. Such simulators can be used for a variety of investigations, particularly in relation to the design of artificial heart valves as described by Bellhouse [16], Black and Melcher [9], Black [17] and Wright [18].

6. The mechanical strength of bone

The work described in the previous sections has dealt entirely with soft tissue. The principal load-bearing structure of the body is, of course, the skeletal system and the mechanical behaviour of the bones of this system has been the subject of many theoretical and experimental investigations. In some cases, these investigations have

*Autologous fascia lata valves are made from a connective tissue which covers the muscles in the thighs of the patient.

dealt solely with the behaviour of bone as a material and not as it relates to the load-bearing capacity of the skeletal system as a whole. A comprehensive study of this aspect of the mechanical behaviour of bone has been presented by Kummar [19].

Other researchers have studied the load-bearing characteristics of bone in order to design more efficient devices for repairing fractures. Work of this nature in relation to the repair of fractures of the neck of the femur has been described by Brown [20] and Hirsch [21].

It is not possible in this present review article to describe in detail the above research. However, one particular investigation undertaken by Frankel and Burstein [22] which involved the testing of long bone under conditions of combined torsional and axial load deserves some comment. The practice of removing small pieces of a patient's own tibia for use in bone grafting in another site is a common orthopaedic surgical procedure. The effect of removing parts of the tibia was to reduce the torsional strength of this bone. This was found to result in fracture of the altered section of tibia under load conditions that would not usually have been considered traumatic.

In their investigation, Frankel and Burstein compared the strengths of various specimens of tibia, both intact and with pieces removed from the cross-section. The three forms of alteration to the cross-section are shown diagrammatically in Fig. 10. Researchers familiar with the problems of the torsional resistance of open and closed structural sections will readily appreciate the complexity of the stress analysis involved in this situation.

The tibia specimens were subjected to combined torsional and axial load in a specially designed test rig as shown in Fig. 11. The results obtained from this investigation indicated that the removal of portions of the bone produced a marked reduction in localized torsional resistance. However, the torsional stiffness of the whole bone was not significantly reduced nor was its axial load-bearing capacity. The main restriction imposed by this procedure of bone removal was that the allowable angular deformation of the bone was greatly reduced.

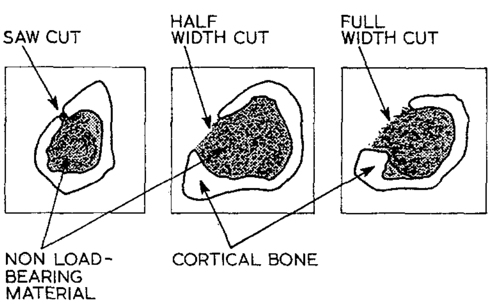

Figure 10 Details of the cross-sections of tibias used in combined axial and torsional loading tests. The three different extents of cortical bone removal are clearly shown.

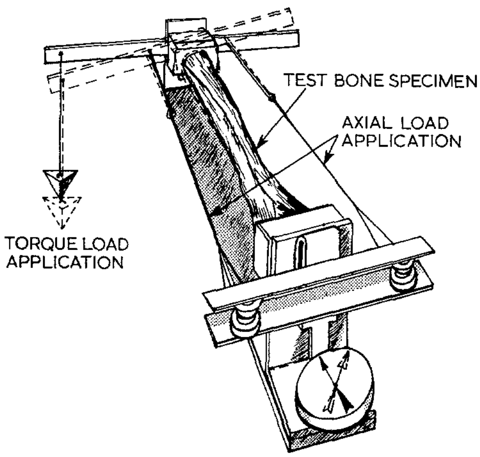

Figure 11 Test rig used for applying combined axial and torsional loads to tibia specimens.

This particular investigation indicates clearly how a biomedical problem can be studied using test methods and forms of analysis already well established in engineering methodology.

7. Conclusions

In this paper, I have tried to present a brief overview of some aspects of the mechanical behaviour of the body's tissues. The subject matter discussed forms a very small part of the work which has been done and indeed is continuing in this field of research. The fact that many other relevant investigations have not been referred to in no way implies any criticism of that work. A collection of several textbooks would be required if this subject were to be treated comprehensively.

At the outset, it was noted that the value of research on the mechanical behaviour of human tissue should not be merely academic. It is to be hoped that the examples quoted have illustrated how the results of this particular kind of research are used to further the efficiency and viability of medical therapy.

Working in this field requires the closest of collaboration between medical and non-medical scientists. This situation involves overcoming the problems of language and, to some extent, the differences in attitude of the life and non-life scientist. However, to judge from research results already published in this field, the above problems are obviously not insurmountable. It is to be hoped that, with the rapid progress now being made in medical technology and the associated sophistication of analytical techniques, more materials scientists and engineers will participate in this form of multi-disciplinary research and development.

References

1. G. A. BORELLI, "De Motu Animalum" (1680).
2. A. K. LANGER, *S.B. Akad. Wiss. Wien.* **44** (1861) 19.
3. R. M. KENEDI, T. GIBSON, and C. H. DALY, "Biomechanics and Related Bio-Engineering Topics", ed. R. M. Kenedi (Pergamon, London, 1965) pp. 147-158.
4. J. E. CRAIK and I. R. R. MCNEIL, *ibid*, pp. 159-164.
5. J. D. C. CRISP. "Biomechanics – Its Foundations and Objectives", ed. Y-C. B. Fung, N. Perrone and M. Anliker (Prentice-Hall, Englewood Cliffs N.J, 1972) pp. 141-179.
6. Y-C. B. FUNG, *ibid*, pp. 181-208.
7. M. ABRAHAMS and T. C. DUGGAN, "Biomechanics and Related Bio-Engineering Topics", ed. R. M. Kenedi (Pergamon, London, 1965) pp. 285-300.
8. R. M. KENEDI, T. GIBSON, and M. ABRAHAMS, *Human Factors* (October 1963) 525.
9. M. M. BLACK and D. H. MELCHER, "Developments in Biomedical Engineering", ed. M. M. Black (Sussex University Press, London, 1972) pp. 52-70.
10. K. ATSUMI and Y. SAKURAI, "Modern Trends in Biomechanics, Vol. 1", ed. D. C. Simpson (Butterworths, London, 1970) pp. 81-120.
11. N. RASHEVSKY, "Some Medical Aspects of Mathematical Biology' (Charles C. Thomas, Springfield, 1964).
12. J. W. REMINGTON, C. R. NOBACK, W. F. HAMILTON and J. J. GOLD, *Amer. J. Physiol.* **153** (1948) 298-308.
13. D. A. MCDONALD, "Blood Flow in Arteries" (Williams and Wilkins, Baltimore, 1960).
14. M. ANLIKER, "Biomechanics—Its Foundations and Objectives", ed. Y-C. B. Fung, N. Perrone and M. Anliker (Prentice-Hall, 1972) pp. 337-379.
15. E. D. MUNDTH, J. E. C. WRIGHT, and W. G. AUSTEN, "Current Topics in Surgical Research – Vol. 3" (Academic Press, New York, 1971) pp. 67-75.
16. B. J. BELLHOUSE, "Developments in Biomedical Engineering", ed. M. M. Black (Sussex University Press, London, 1972) pp. 71-82.
17. M. M. BLACK, "Perspectives in Biomedical Engineering", ed. R. M. Kenedi (to be published by Macmillan, London, 1973).
18. J. T. M. WRIGHT, "Developments in Biomedical Engineering", ed. M. M. Black (Sussex University Press, 1972) pp. 33-51.
19. B. K. F. KUMMAR, "Biomechanics – its Foundation and Objectives", ed. Y-C. B. Fung, N. Perrone and M. Anliker (Prentice-Hall, 1972) pp. 237-271.
20. J. T. BROWN, "Biomechanics and Related Bio-Engineering Topics", ed. R. M. Kenedi (Pergamon, London, 1965) pp. 325-339.
21. C. HIRSCH, *ibid*, pp. 341-357.
22. V. H. FRANKEL, and A. H. BURSTEIN, *ibid*, pp. 381-396.

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