PHYS. MED. BIOL., 1975, VOL. 20, NO. 5, 699–717. © 1975

Review Article

Tissue Mechanics

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This review was completed in February 1975

Contents

1. Preface

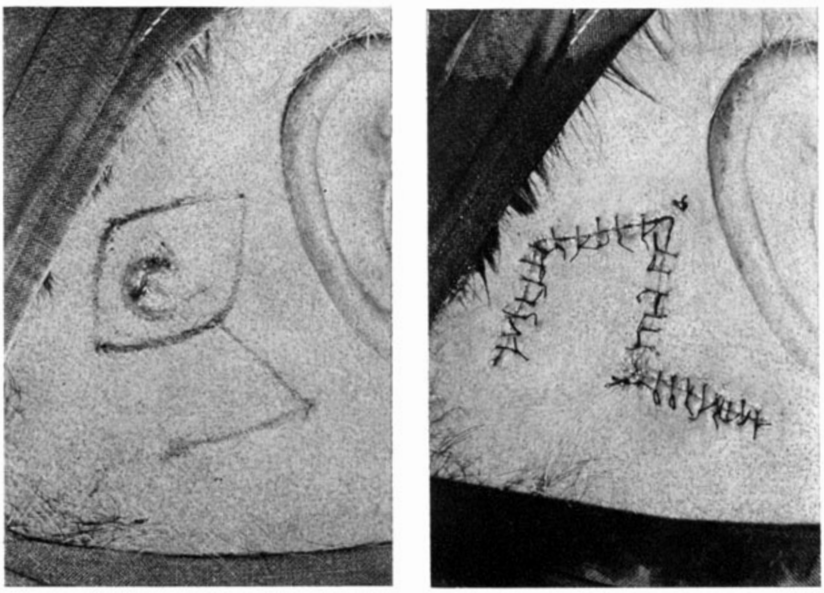
The human body, from an engineering point of view, is a load-transmitting mechanism. It is subjected to self-originated and/or environmentally-imposed actions (Abrahams 1965, Paul 1967). These in turn manifest themselves in actions on tissue, and the tissue's capability to withstand and transmit the corresponding loads through deforming appropriately can be characterized in health and disease. Knowledge of such 'load-deformation' relations of the tissues can serve as a basis for diagnosis, for the development of techniques of therapy (in reconstructive surgery for example) and in general for monitoring treatment progress (Jansen 1955, Vlasblom 1967, Finlay 1970).

The literature of tissue mechanics in its biological, biochemical, mechanical and other aspects is intimidating in its extent (see appended Bibliography). Despite this it is still far easier to list knowledge gaps which exist and work that requires to be done than to identify any achievement commensurate with the research effort expended and which can truly be claimed as relevant to the human patient.

The authors—believers in pragmatic bioengineering—attempt in this paper to explore the reasons for this. Their competence is limited to the medical and mechanical aspects and the paper is presented within such limitations. It will, however, emerge that even such a limited exploration uncovers the need for a reconsideration of research orientation so as to re-establish the clinical problems of the patient, both as the starting point and as the ultimate applied aim of all research enquiries.

2. Basic concepts

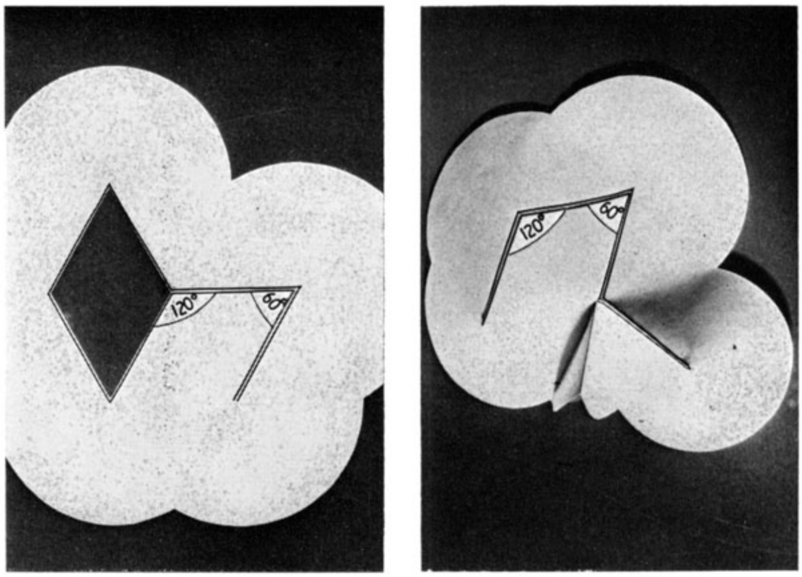
The rationale for the initiation and development of tissue mechanics springs from the concept that the human body, as already stated, is a load-transmitting mechanism. There are very many facets of this, but possibly the most visually obvious ones are found in the procedures of reconstructive surgery (Gibson and Kenedi 1967). Fig. 1 shows an example of this. To repair a skin defect the



(a) (b)

Fig. 1. (a) Procedure to remove rodent ulcer from behind the ear. Proposed incision lines marked on skin and rhomboid, containing ulcer, to be excised. (b) Immediate post-operative appearance. Owing to the extensible nature of the tissues, large deformations occur in the plane of the skin. The procedure is accommodated with acceptable tissue distortion.

reconstructive surgeon removes an area of full-thickness skin and then covers this void by a topographical rearrangement of the surrounding skin (Limberg 1929, MacGregor 1957, Gibson 1965). It should be noted that there is no addition of tissue and that removed is compensated for in part by the 'mobility' of the skin and in part by its deformability under the loads imposed on it by the surgical manipulation. It is obvious (in an over-simplified kind of way) that this procedure may be regarded as a type of engineering construction in the 'unconventional' material of human tissue. Initial model analyses were carried out (Limberg 1929) using inextensible paper as shown in fig. 2. The need for a more realistic analysis was one of the many triggers that brought engineers into this area.



(*a*) (*b*)

Fig. 2. (*a*) Paper model with incision lines and rhomboid area of excision. (*b*) Paper model showing flap transposition and the local distortions produced in this inextensible medium.

Initially the engineering approach was of blunderbuss type crudity. It amounted to carrying out tissue load–deformation tests on the macroscale (Kenedi, Gibson and Daly 1965) which, for convenience and ease, were performed in tension. Fig. 3 shows such a load–deformation test in uniaxial tension on human skin. The features to be noted here are that the curve shape is concave to the load axis, showing a decrease in deformation with increase in

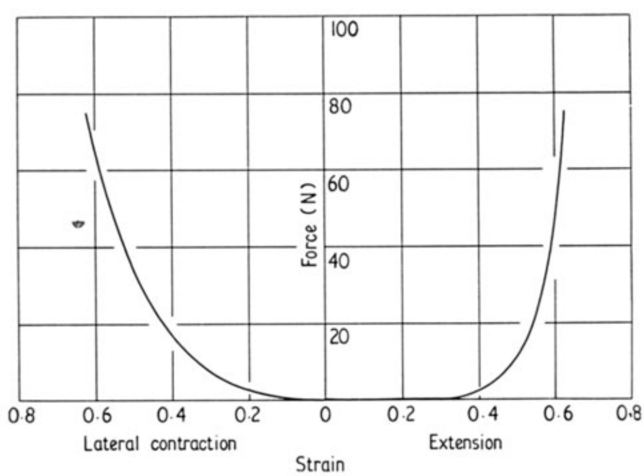


Fig. 3. Force–deformation relations for excised human skin in uniaxial tension. Large extensional strain is accompanied by contraction of a similar magnitude.

load, and the contractions at a right angle to the applied load (the so-called Poisson effect) are comparable in magnitude to the direct extensions. Fig. 4 shows similar load–deformation curves for a variety of tissues and fig. 5 illustrates the progressive modifications consequent on cyclic load application. Incidentally, such characteristics are now considered to have been reliably established as common to almost all non-calcified tissues (Nachemson and

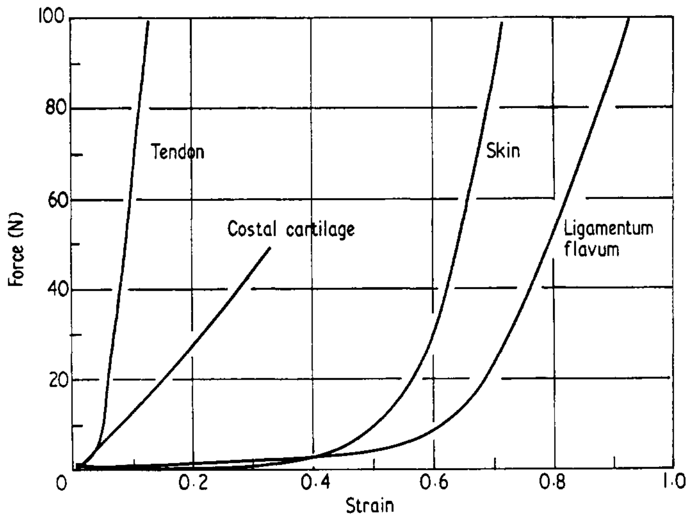


Fig. 4. Force–deformation relations for a selection of excised human tissues tested in uniaxial tension. Apart from costal cartilage which is sensibly linear, other tissues display an initial lax response with a gradual transition into a stiffer, nearly linear response. The magnitude of strain at the transition is the principal variable.

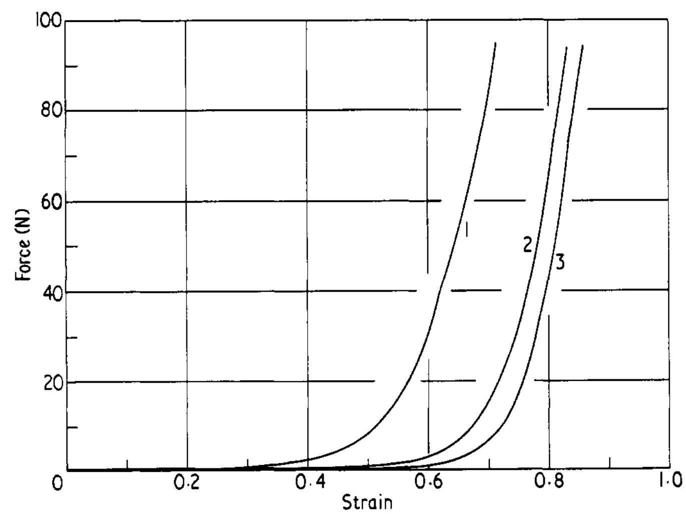


Fig. 5. Repeated extension cycling of skin to a prescribed force results in a stabilization of the relation between force and deformation. This process, often termed 'preconditioning', is commonly employed to yield stable characteristics relatively independent of previous manipulation.

Evans 1967, Stromberg and Wiederhielm 1969, Bergel (1972) and Fung (1972), Viidik 1973). The assessment of uniaxial load–deformation behaviour obviously represents only part of the tissues' response.

One of the primary additional influences is the time dependence of all macrocharacteristics. As an example, tensile tests on human tendon (fig. 6) at varying strain-rates show significant time-rate dependence.

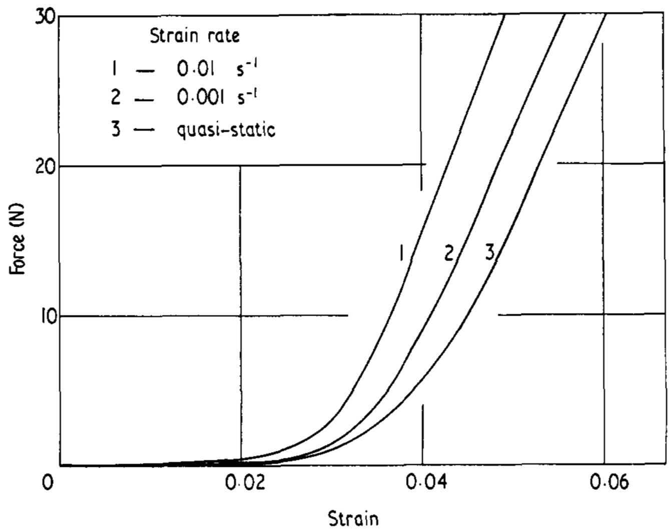


Fig. 6. Strain rate dependence of the force–deformation characteristics of human tendon in tension. Curve 1 corresponds to the lower strain rate experienced by tendon during normal walking.

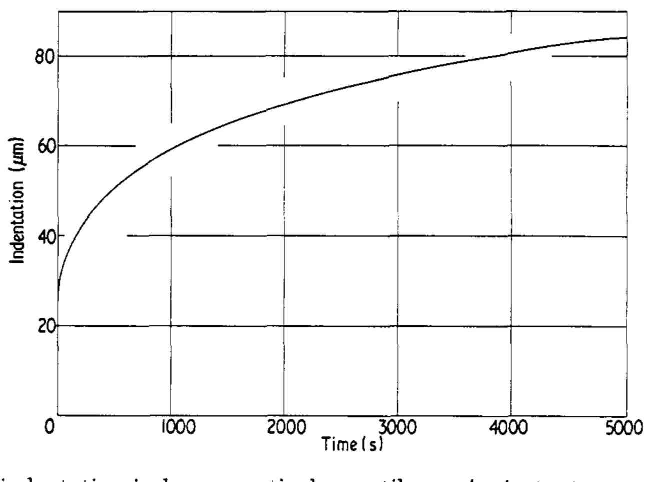


Fig. 7. Creep indentation in human articular cartilage. An instantaneous response precedes time-dependent deformation of a similar magnitude.

Creep and stress relaxation effects also occur (figs 7 and 8). This in turn implies that comparison of test results is only relevant if their time rates are controlled and known. It is unfortunate that a large portion of the literature reporting careful and well meant work is of questionable value because of the disregard of this time factor.

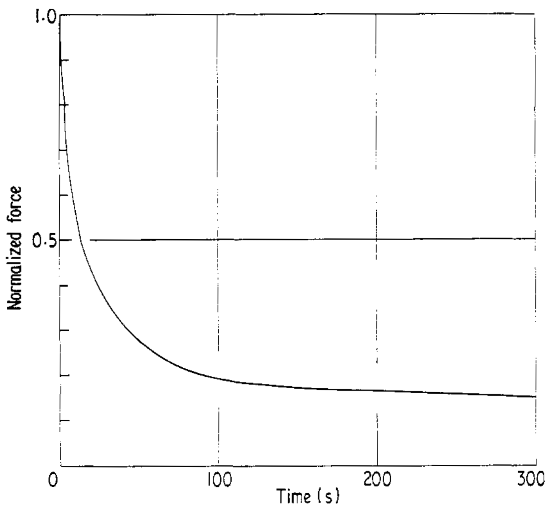


Fig. 8. Relaxation of indentation force in articular cartilage from the rabbit. The normalized equilibrium force of approximately 0·15 is attained within minutes of relaxation commencing.

The macrolevel load–deformation investigations initiated by uniaxial types of testing have been extended to explore, for example, the effect of biaxial deformations in the plane of the skin, human and animal (Vlasblom 1967, Grahame 1969, Hilderbrandt, Fukaya and Martin 1969a, Finlay 1970, Lanir and Fung 1974).

These show in a qualitative sense the same kind of overall relationships that are seen to obtain under uniaxial conditions (Kenedi *et al.* 1965, Veronda and Westmann 1970, Fung 1972, Barbenel, Evans and Finlay 1973).

Biaxial tests in particular demonstrate very clearly the difficulties encountered when attempting to apply the 'continuum' concept to biological tissues. There is disagreement over whether these tissues are compressible (Kenedi *et al.* 1965, Veronda and Westmann 1970) or sensibly incompressible (Fung 1967, Blatz, Chu and Wayland 1969, Hilderbrandt, Fukaya and Martin 1969a). To complicate matters further, significant anisotropy has been demonstrated both *in vivo* and *in vitro* for a wide variety of tissues.

Consideration of the full complexity of the problem requires elucidation of at least two other aspects; the micro structure of tissue and its function in the particular animal or human body of which it forms part.

3. Structure and function of tissue

In the foregoing discussion, characterization was based on experimentally observed features on the macroscale without the microstructure of tissue and its components being explored.

Considering human skin as an example, it is seen that it has a multicomponent microstructure, the basic elements of which are five intertwined networks of collagen, elastic and nerve fibres, small blood vessels and lymphatics, covered by a layer of partially keratinized epithelium and transfixed at intervals by hairs and the ducts of sweat glands (fig. 9). The networks are surrounded by

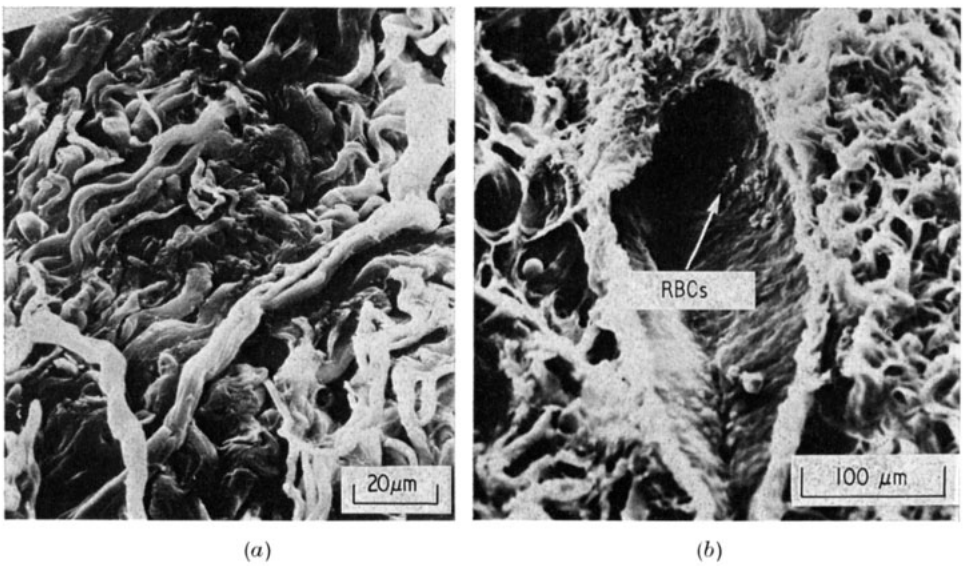


Fig. 9. (*a*) Dermal collagen fibre weave in the skin of a child. The fibres are convoluted and unconnected structures. (*b*) An obliquely cut blood vessel in the mid-dermal region. A cluster of red blood cells (RBCs) can be seen lying against the wall of the vessel.

interstitial fluid containing a varying amount of mucopolysaccharide ground substance and all are mobile in this semi-fluid environment (Gibson, Kenedi and Craik 1965, Finlay 1969). In the palm and sole, the thick, keratinous outer layer probably plays a major role in defining the mechanical properties of the skin (Harkness 1971). Elsewhere only the fibrous networks of collagen and elastin, the interstitial fluid and, depending on the type of loading, the relevant subcutaneous tissues appear to need serious consideration. It should be remembered however that the blood vessels, nerves and lymphatics are intimately enmeshed with these fibrous networks (Brown 1973) and blanching, pain and oedema can result when the fibrous pattern is deformed. The effect of hairs passing through the networks is uncertain. The short lengths which lie in the dermis may have sufficient rigidity to act as pegs and thus reduce fibre network deformation (Brown 1973).

It must be particularly emphasized that all tissues are living organs, and thus in addition to the time dependence of the mechanical effects already demonstrated, biological effects can also occur during tests modifying the properties being measured.

All connective tissues exhibit complexity of the micro-architectural arrangements of their components in greater or lesser degree. They can be classified as proposed by Evans (1973) into tissues characterized by 'dominant' and 'interactive' components. While it is probably a gross oversimplification to say that the structural complexity of a given connective tissue is directly related to its diversity of function in the body, it is generally true to say that the simpler the structure the more circumscribed is the apparent functional requirement.

Thus tendon, having a more unidirectionally oriented function than say skin, also has a simpler structure. For most tendons this consists primarily of helically coiled collagen bundles (Lerch 1950, Keller 1951, Künke 1962, Barbenel, Evans and Gibson 1971) and these can in consequence be classified as unifunctional, one-component 'dominant' tissues.

In attempting to classify tissues in this fashion it is needful to recall again the ever-present time dependence of the tissue constituents. Because of this, the microstructure, the interaction of the tissue components and the resulting overall response of a given tissue, is certain to change from instant to instant even though apparent macrostability of deformation, for example, is maintained.

It is also necessary that understanding be achieved of the load–deformation relationships *in vivo* and particularly in what has been termed the 'physiological' range. Evans (1973) suggested, and his co-workers subscribe to this concept, that it is in this physiological range that his classification of dominant as opposed to interactive component tissues can be used to most advantage.

Examples of human and animal tissue grouping in terms of Evans' classification are as follows:

(1) Tissues with a relatively simple structure and one dominant component: tendon (dominated by collagen), ligamentum nuchae (elastin), ligamentum flavum (elastin) (Elliot 1965, Wood 1954, Nachemson and Evans 1967).

(2) Tissues with a relatively complex micro-architecture yet whose mechanical response is still dominated by one component: elastic fibro cartilage (elastin), costal and nasal cartilage (ground substance), peripheral nerve trunking and fascia (collagen) (Gratz 1931, Abrahams and Duggan 1965, Fry 1966, Sunderland 1968, Millington, Gibson, Evans and Barbenel 1971).

(3) Tissues with complex structure and interactive component behaviour: aorta (elastin, collagen and muscle) (Bergel 1972) and one of the most complex, skin, in which elastin collagen and ground substance interact to varying degrees at virtually every stage of loading.

The importance of the interaction of the mechanical response and of certain physiological limits cannot be overemphasized. One of these physiological limits is related to the tissue's blood supply. If this is inhibited consequent on load and/or deformation becoming excessive (Hickmann, Lindan, Reswick and Scanlan 1966), the tolerance threshold of the tissue can be reached or exceeded.

Maintaining such conditions can lead to tissue damage or even tissue death. Although regarded clinically as of major importance, viability studies of tissue concerned with establishing such tolerance thresholds are few in number and of doubtful applicability. The recent work of McGregor (1975) exploring the major 'axes' of the superficial blood supply of human skin is of particular significance. It appears that there is a likely correlation between such major lines of blood supply and biomechanical characteristics such as the lines of anisotropy in skin (Kenedi, Evans and Barbenel 1975).

4. Biomechanical characterization and modelling of tissue

The authors might be forgiven if they comment with a degree of asperity on efforts (including their own) which have been deployed in this area to date. Attempts at modelling exhibit intellectual ingenuity of a high order, are academically entertaining and on the whole are as yet useless from the applied point of view.

Biomechanical characterization and modelling cover basic ideas and other explorations, some of which have added significantly to knowledge and understanding of tissue behaviour (Hilderbrandt *et al.* 1969b, Stromberg and Wiederhielm 1969, Crisp 1972, Diamant, Keller, Baer, Litt and Arridge 1972, Fung 1972, Barbenel *et al.* 1973, Viidik 1973).

Modelling is being undertaken in two primary areas. Firstly, in the development of artificial replacements for tissues. Here characterization of natural tissue behaviour is needful so as to be able to specify realistically the artificial replacement or synthetic simulant. Secondly, tissue modelling is considered in relation to the diagnostic area. Attention here has been concentrated on skin for the reason that it is the most accessible tissue in the human body and also that, in a number of instances, it tends to reflect the state of health of the body as a whole (Chvapil 1967). Thus, for example, in rheumatoid conditions, the collagen of skin is affected. The progress of the disease and the effects of therapy can then be reflected in the biomechanical characteristics of the patient's skin.

Modelling itself has taken three forms, the first of which uses artificial materials. As some tissues were initially conceived as having elastomeric characteristics, materials such as silicone rubbers have been employed. The authors consider this simplistic approach unlikely to be successful and regard it now as only of historic interest in the development of tissue mechanics.

Secondly, there has been the mathematical approach to modelling. This has taken a variety of forms. One is the use of spring and dashpot rheological models. While these can effectively represent, for example, the overall uniaxial load–deformation responses of tendon, their use has also given rise to incorrect attempts at identifying the mathematical components of the rheological model with physical ones of the tissue.

The third approach has been the use of continuum formulations in uni- and multi-axial situations (Blatz *et al.* 1969, Hilderbrandt *et al.* 1969a, b, Fung 1972, Barbenel *et al.* 1973). Some of these works are elegant and sound.

Virtually every one of them, however, disregards features such as the influence of the load–deformation history of the tissue prior to the particular instance considered, anisotropy and time dependence. Any attempt at a more embracing theory concerning such factors leads to intractable mathematical and experimental complexity.

5. Clinical applications

As already mentioned, two of the major areas to which tissue biomechanics is at present oriented are the specification of the required characteristics of tissue replacements and implants and the evaluation of the change of biomechanical characteristics of tissues in health and in disease so as to provide diagnostic indices.

To extend these areas of clinical applications and to introduce a somewhat unconventional illustration of biomechanical considerations, the example chosen for consideration is that of wound 'design'. This is feasible in operation wounds such as incisions, excisions, flaps, etc., which are forms of electively controlled surgery.

To date the total effort directed to developing rational bases for such wound or flap design is disconcertingly small. Research in this area also tends to be fragmented and considers usually one aspect only, thus providing at best 'partial solution' pointers. Skin flap design has received most attention and solutions have been based initially on assumptions that tissue is inextensible and that mechanical characteristics are the same in every direction (Limberg 1929, McGregor 1957, Devlin 1965, Furnas 1965).

A more sophisticated approach was initiated by Stark (1971). He firstly showed that Langer's lines (Langer 1862, Ridge and Wright 1966, Gibson, Stark and Evans 1969) in human skin were lines of minimal extensibility and that operation incisions at right angles to these lines produce minimum wound opening (fig. 10). He then expanded his work into considering design of excisions and explored in particular the optimal oval excisable so as still to leave a closable wound. Accepted surgical practice tends to use a length to breadth ratio for such an excision of about 4 to 1. Stark has shown (confirmed in selected operations) that the length to breadth ratio is dependent on local biomechanical characteristics of the skin (a composite of the physiologically permissible extensibility and mobility) adjacent to the excision. In a case he quoted, the designed optimal length to breadth ratio was 1·6 to 1. This provided not only a closable wound but avoided also the 'dog's ears' deformity that tends to obtain when closing a wound produced by tissue excision. The major difficulties in extending Stark's concepts to more complex flat design arise from the present inability to analyse effectively large deformations of tissue, from lack of knowledge of skin mobility *in vivo* and finally from a lack of basic understanding of the mechanism, nature and extent of the tissue's blood supply.

Thus one can see that despite the extensive wound healing and repair and other literature, practically applicable data are still very much of the empirical kind. Clinically usable information is provided in particular by practising surgeons rather than by the wound repair researchers. The form of derivation of such data shows at times innovative ingenuity to which biomechanicians have also contributed.

A recent example of this is a modification to reduction mammaplasty introduced by G. Morgan (Canniesburn Hospital), R. Wilkinson and D. S. Ross (University of Strathclyde).

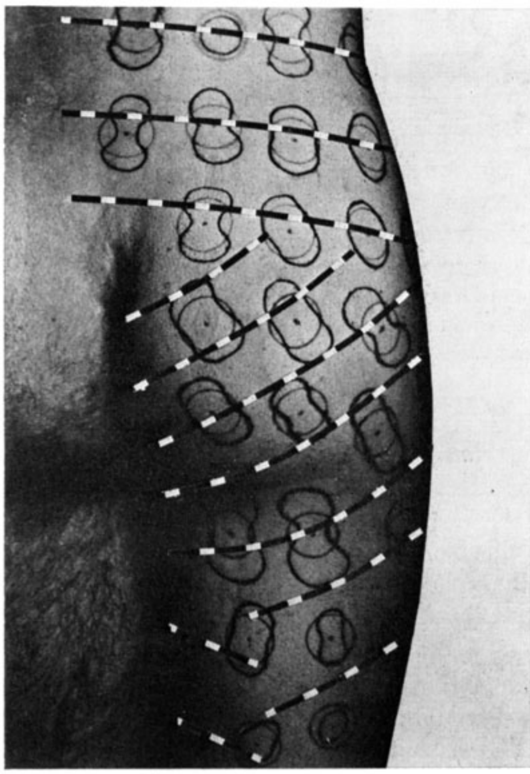


Fig. 10. Anisotropy of intact human skin. The directional variations in strain are plotted at a number of sites on the back of a male subject aged 25 years. These strains obtain on application of a constant uniaxial tensile load corresponding to the transition between lax and stiff response (see fig. 4). The circles represent a strain of 0·20. The superimposed broken lines correspond to Langer's lines (see text).

In cases of enlargement or malformation of breasts, operations are undertaken to reshape these. Morgan had the idea that in this kind of operation it would be an estimable additional aim to produce a result which would approach the 'ideal' shaped breasts. For this reason, through a consensus of the cognoscenti, a human model having the ideal proportions was 'elected'. The biomechanicians took casts of the breasts and characterized them in an engineering manner, that is, they obtained their surface areas and contours, then developed these into a plane shape so as to provide an effective model template for the 'ideal shape' proportions. Initially only intuitive allowance based on clinical experience has been made for the time-dependent post-operative tissue deformation. Now a significant series of patients who have undergone this kind of reconstructive surgery are under long-term survey. From this it is hoped to obtain reliable data and appropriate 'correction' factors applicable to the template at operation, so as to make the long-term result approach that initially designed.

6. Summary

It will be appreciated, and it is immediately apparent from the classified annotated bibliography appended, that the literature of tissue mechanics is remarkably extensive and that its aims are being pursued in an expanding and increasingly active fashion by many workers world-wide. It is unfortunately also obvious that very little of the work at present pursued is applicable in a clinical sense. This is particularly so regarding the bulk of the biomechanical animal work, which forms such a substantial portion of the literature and yet is virtually inapplicable to the human case. This is firstly because biomechanical characteristics such as anisotropy, time dependence, preconditioning, etc., in animals and in humans are not comparable. Secondly, attempts to obtain quantitatively-consistent results in the animal experiments usually lead to refinements of experimental control to the point where the results are virtually inapplicable to any practical situation even in the animal context itself.

Rethinking of research orientation in the tissue mechanics area is therefore necessary. This in the authors' view should concentrate on exploring specifically human tissue, its deformability, its mobility and physiological viability. Such data should then be applied to the problems of the human patient in diagnosis and in therapy.

Some of the investigations included in the paper have been carried out in the laboratories of the Bioengineering Unit, University of Strathclyde (Director, Professor R. M. Kenedi). The work is generally under the clinical direction of Professor T. Gibson and that reported here has been carried out (in addition to the authors) by Drs. P. F. Millington, J. A. Brown, G. R. Fernie, J. B. Finlay, F. Gibson, R. Mulholland, J. F. North, H. L. Stark, R. Wilkinson and Mr. L. Juhasz.

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