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The mechanical Design of Skin - towards the Development of new Materials

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Abstract

Skins - the soft outer coverings of many vertebrate and invertebrate animals, are made of collagen fibres arranged in more or less ordered arrays. Skins containing well ordered arrays such as the helically wound collagen fibres of worms and some fish, are probably relatively easy to model using simple geometrical models. Skins containing more randomly arranged collagen fibres are more difficult to model. In fact no credible model yet exists. We consider that these skins present a complex problem more likely to be solved using modern computer-based methods capable of calculating the mechanical contribution of large numbers of individual fibres.

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[1] The partnership of authors in this paper follows directly from Drs Vincent and Topping meeting at the SFB 230 conference in 1989.

Introduction

There are many soft biological composite materials such as skin, blood vessels, tendons and other animal connective tissues which are composed of an array of collagen (and frequently elastin) fibres in a protein-polysaccharide matrix (figure 1 shows some examples). In addition the collagen fibres may be straight, crimped or convoluted to various degrees and arranged in varying degrees of apparent randomness. In human skin there is a limited degree of orientation, reflected in Langer's lines; in earthworms, sharks and nematodes there is a very high degree of orientation which can be directly related to transmission of force by the skin, although imperfect knowledge of the three-dimensional effects on such a material (e.g. the amount by which it gets thinner as it stretches) must inevitably limit our understanding of its function and the resulting implications for the animal. Other animals such as coelenterates (both sessile and mobile) have a complex system of fibres in the body wall which has not been properly or adequately analysed (Wainwright *et al.* 1976; Vincent 1991). In the walls of blood vessels and the body wall of sea anemones, the mechanical properties have been shown to depend strongly on the orientation of the collagen fibres in the network; it has been shown that the collagen fibres are substantially reorientated on extension of the tissue, so producing significant strain-dependent changes in their mechanical properties (Bigi *et al.* 1981, 1985). Diamant *et al.* (1972) have proposed a model for tensile behaviour of crimped collagen fibres (the "elastica" model) which can be combined with the mechanical properties of a fibre network (Krenchel 1964) to give a workable model for a real network of fibres (Purslow 1989). The measurement of orientation of fibres is possible using x-ray diffraction (Purslow 1989) and optical (polarisation) microscopical techniques.

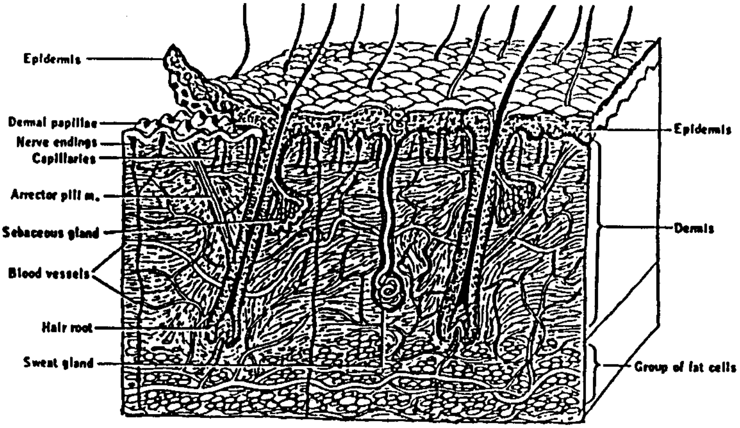


Fig. 1a: Section of mammalian skin to show hair, glands and accessory structures.

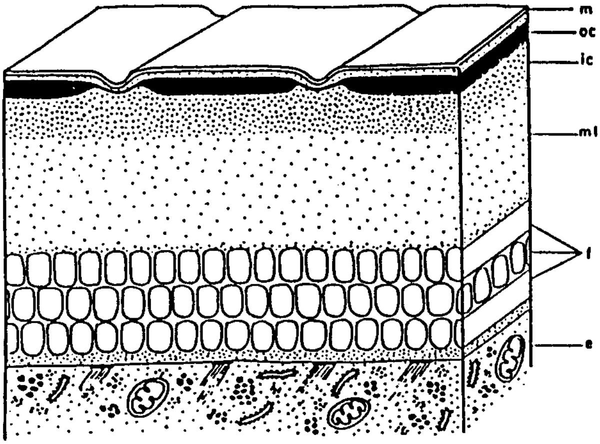


Fig. 1b: Diagram of outer covering (cuticle or skin) of a roundworm in which the basal layer contains large collagen fibres. Abbreviations: e, epidermis; f, fibre layers; ic, inner cortex; m, membrane-like layer; ml, median layer; oc, outer cortex.

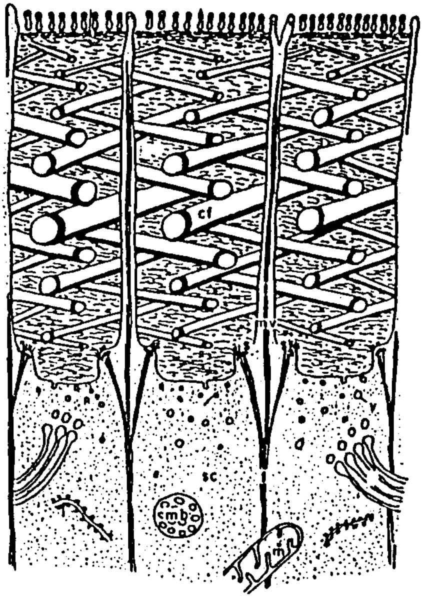


Fig. 1c: Diagram of the outer covering (cuticle or skin) of an earthworm. The collagen fibres (cf) are arranged in a crossed lamellar fashion in the upper half of the diagram. Microvilli (m) extend outwards through the skin from the supporting cells (sc) and attach to the skin via tonofilaments (t) and shoulders (s) with attached microfilaments.

The important characteristics of these tissues are heterogeneity, anisotropy and the mobility of the fibres. Depending on the resistance to crack propagation and degree of initial orientation of the fibres, these materials can be stretched to high (10% to 50%)

strains which are associated with an elastic modulus increasing with deformation giving a J-shaped stress-strain curve (figure 2). This curve has many interesting characteristics, not least of which is that it seems to be associated with high toughness (Mai & Atkins 1989), perhaps an underlying reason for using such an arrangement of fibres in an outer covering. Other reasons for wanting to understand the basis of the stress-strain characteristics are both medical (plastic surgery; design of prostheses; replacement skin for burns victims, etc.) and engineering (such soft tissues are remarkably widespread in the animal world and so may represent a useful, safe and efficient way of containing materials, especially under pressure, which could be modelled if we knew the important characteristics).

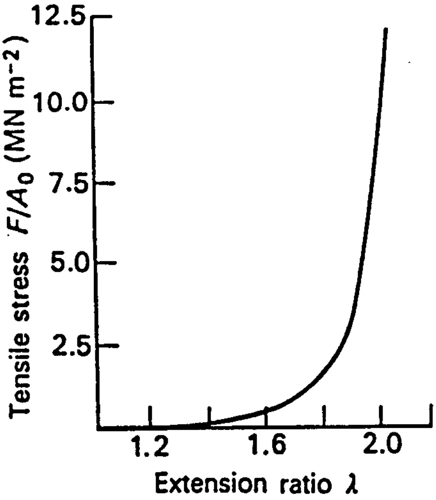


Fig. 2: Stress-extension ratio curve for cat skin loaded uniaxially to failure.

Current Models

There are many papers in which the mechanical properties of skin (and other soft tissues) are reported and analysed. Fibre-wound pressure vessels are fairly common, there being many worms which have their fibres wound at close to the "magic angle" of about 54 (one of the zoological origins of these ideas is summarised in Clark 1964. See also Wainwright *et al.* 1976). However, this model is very simple, with all fibres changing their orientation uniformly and simultaneously. Most soft tissues are much more complex with semi-randomly orientated fibres arranged in a 3-dimensional network.

On the whole, models have tended to predict only unidirectional stress-strain behaviour of these tissues (i.e. they can describe only elastic, albeit high strain, reversible deformations), paying little or no attention to strength and fracture properties. These models have progressed from a simple power law (Ridge & Wright 1964) via constitutive equations of greater complexity (Shoemaker *et al.* 1986) and less credibility (Demiray 1988, which starts with the statement "Assuming the arterial wall is homogeneous, incompressible, isotropic and elastic .. ", all of which it is not), to a more structurally-based approach which can describe both re-orientation with strain and two-dimensional strain (Lanir 1979) and one in which the parameters of a basically constitutive model can be directly related to the mechanical properties and dimensions of the collagen fibrils (Manschot & Brakkee 1986). Jeronimidis & Vincent (1984) reviewed some of the problems inherent in applying composite theory to such tissues, including an analysis of Poisson ratio effects in soft collagenous tissues which has since been followed by a short experimental study (Lees *et al.* 1991). Aspden (1988) developed this approach and considered a fibre composite model based totally on simple reorientation and stretching of fibres. He applied it to changes in the mechanical properties of the human cervix during pregnancy.

A basic problem with all these models is that they require information on basic "matrix" and "fibre" properties which are not directly measurable or have not been measured directly but inferred from measurements of the composites themselves. For instance the accepted value for the Young modulus of collagen fibres is only 1.5 GPa or so. This value is typical of amorphous polymers, but one would expect a higher value from what is known to be a relatively crystalline material in which the tensile stress is borne directly by the main-chain covalent bonds. Silk (another fibrous protein, highly crystalline but lacking the matrix material found in tendon) has a modulus nearer 100 GPa. It seems likely that the modulus of collagen is much higher, though the highest reported value we have been able to find is 10 GPa (Chien & Chang 1972).

None of these models is satisfactory for one or more of the following reasons:

1. generally, only uniaxial strains are modelled;

2. the model is statistical in its approach, eliminating the heterogeneity of skin by assuming it is a continuum with a constant Poisson ratio. This may imply lack of understanding of the "representative volume" of the materials;

3. the mathematical problems of high strains are often ignored;

4. the effects of imperfections or local variations in strain as might occur in the vicinity of a cut (Purslow *et al.* 1984, Broom 1984) are not considered and cannot be modelled;

5. the failure and fracture properties are almost totally ignored;

6. no account is taken of prestrain.

The effect of these analytical shortcomings is that there is an almost total lack of usable tools for designing, or predicting the properties of, these complex materials. This is especially apparent with aspects of fracture and failure (4 and 5 above).

Proposed Approach

In order to model the properties of skin, the following factors have to be accounted for:

- stress-strain and stiffness properties of all fibres
- relevant properties of the matrix
- re-orientation of whole fibres and segments of fibres during extension
- uneven strain fields in the vicinity of cuts and other imperfections
- high extensions
- gradual controlled breakdown of structure as failure proceeds (i.e. localisation and evolution of damage)
- visco-elastic properties of the matrix and fibres
- edge and end effects on the specimen (these should be apparent from correct modelling of biaxial stress-strain behaviour)
- Poisson ratio effects
- one-dimensional and two-dimensional strain

We consider that the failure to model skin and other tissues in a convincing manner is due to the relatively simple methods which have been used. We report here the preliminary stages in the development of a novel approach which is basically more complex, using a combination of finite element theory and reorientating nets using advanced computing techniques. It has the following unique strengths:

- It can cope with all combinations of strain in the plane of the sheet
- It needs no extra information about the Poisson ratio
- It can cope with edge effects and imperfections

- It takes any number of different types of fibre with different stress strain and viscoelastic characteristics

- It requires no special mathematics to cope with high strains

- It offers the possibility of generating totally new types of material which will be compliant in bending and extension, will be very tough and able to respond to local imperfections (cuts, perforations) by modifying its structure to resist further damage.

To this extent these novel materials qualify as "intelligent" or responsive to local conditions.

Preliminary Results

Method of Analysis

The structural analysis of skin requires a capablility to model large changes in the structural idealisation which are due to changes to both material and geometric properties. Experience in the analysis of architectural tension structures has shown that it is frequently difficult to obtain satisfactory solutions to such grossly non-linear problems using implicit solution schemes such as Newton or Newton-Raphson approaches. Consequently the explicit dynamic relaxation procedure was adopted for this preliminary study. The dynamic relaxation procedure is an explicit time stepping method for the solution of linear and non-linear static structural analysis problems. In this the ficticious dynamic behaviour of the structure was removed by a "kinetic damping" procedure. The explicit nature of this technique permits the inclusion of many different types of finite elements and "on-off" non-linear material behaviour may be readily accounted for. In addition large displacements may be readily accounted for by using the current geometry of the structure at any instance in time. This permits the analysis of no-compression membrane structures with intertwined cables which has obvious similarities with skin (see above).

Adaptive Procedures

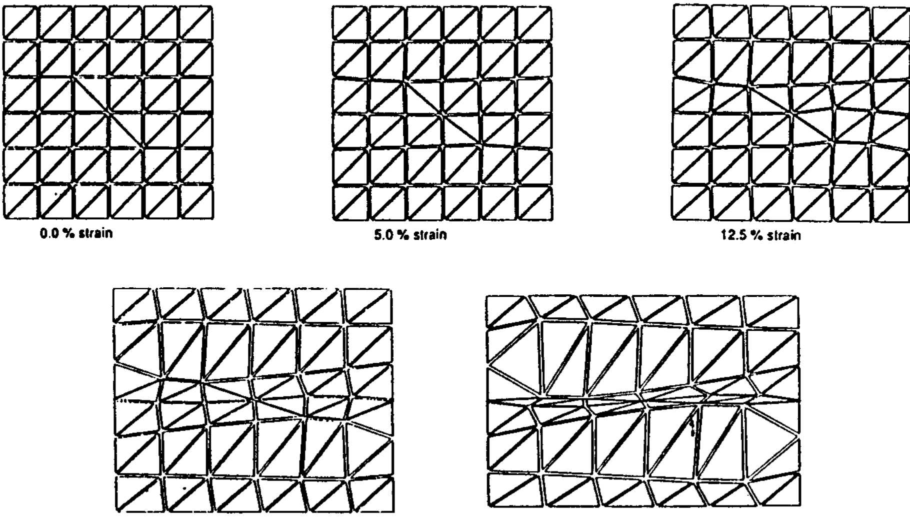
In finite element idealizations based upon simple constant-strain triangular elements the accuracy of the results depends upon the topology of the mesh chosen for the finite element analysis. Adaptive remeshing can predict an efficient mesh by taking into account the domain error for a uniformly graded mesh. An efficient mesh is defined as the one in which the error is equally distributed over the domain. Through adaptive remeshing the domain error is reduced as well as uniformly distributed over the domain using an un-structured mesh generator.

The analyses are carried out until the domain error becomes less than a pre-defined error value (Zienkiewicz & Zhu 1987). The errors were calculated using nodal averaging of element stresses and a Root Mean Square averaging of the local errors in energy over the whole of the domain (Khan & Topping 1991).

Modelling Skin

In the following case study adaptivity and dynamic relaxation have been combined to model skin type material. The skin was modelled using constant stress triangular elements for the ground tissue and a cable type element for the collagen fibre. The modulus of elasticity for the membrane and the cable were taken as 1 kN/mm^2 and 2 GN/mm^2 respectively. As remarked above, this stiffness for the cable is too low if it is to represent collagen. The current model shows little difference if this stiffness is increased by a factor of 6, but we need more believable data for collagen for further modelling. The Poisson ratio for the membrane was kept as 0.49.

Figure 3a shows the rectangular membrane, idealized using a regular grid mesh which was completely restrained on the right and the left boundaries but free to slide horizontally on the upper and the lower boundaries. A meandering cable was placed through the membrane and fixed at its ends on the right and the left boundaries. Figures 3b and 3c show the same membrane after the right hand boundary has been moved to provide strains of 5% and 12.5% respectively. Figures 3d and 3e show that the same membrane with the right boundary further moved to the right to provide the strains of 25% and 50% respectively. It is apparent from these figures that at such large strains the finite element idealisations have become grossly distorted which will result in excessive errors in the analysis.



0.0 % strain

5.0 % strain

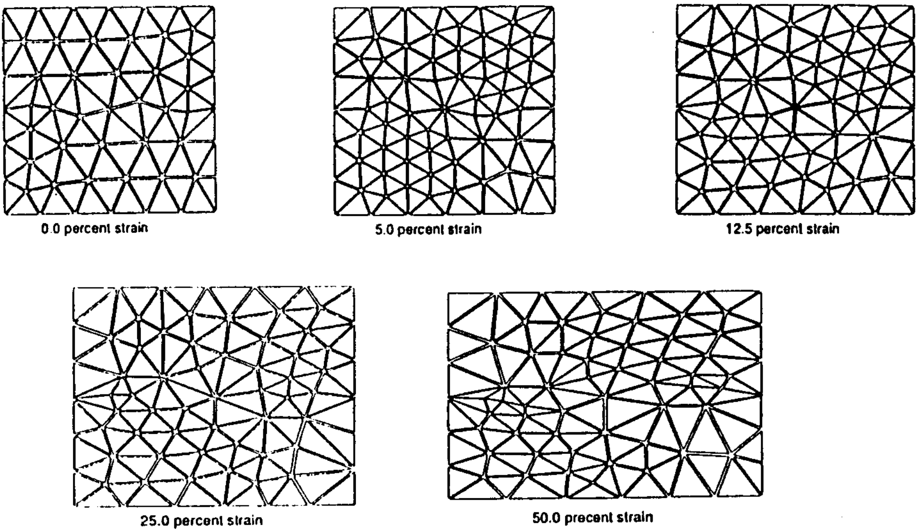
12.5 % strain

25.0 % strain

50.0 % strain

Fig. 3a-e: Extension of a cable in a membrane using 'simple' finite elements.

The complete analysis was undertaken again but this time incorporating the adaptive procedure described above. Figures 4a-4e show the membrane at strains of 0%, 5%, 12.5%, 25.0% and 50.0%. The adaptive procedure which was performed at each stage of analysis ensured a well-conditioned mesh and that the true geometrically non-linear structure was followed.



0.0 percent strain

5.0 percent strain

12.5 percent strain

25.0 percent strain

50.0 precent strain

Fig. 4a-e: Extension of a cable in a membrane using adaptive remeshing of the finite elements, thus reducing distortion of the elements.

Figure 5 shows the force displacement characteristics of the complete membrane idealised using a regular grid and an adaptive un-structured mesh. These two graphs illustrate that an adaptive procedure is required accurately to predict the behaviour of the complete membrane. This is a J-shaped curve which (as noted above) is typical of skin and similar tissues. The precise shape has been shown to vary with different soft tissues, so with the variables which we can incorporate into our model we see no obstacle to producing even more realistic models.

Using this model we shall be able to establish the controlling factors for the mechanical behaviour of skin. The understanding of the criticality of defects in skin-like structures will provide the basis for identifying the materials and/or structural factors which control resistance to fracture. It will be possible to assess the relative importance of several anticipated parameters such as fibre-matrix interactions, ratios of elastic and strength properties of fibre and matrix, local and fibre reorientation by computer modelling and by experiment. Using physical models as well as real systems it will be possible to identify some of the parameters, chemically, biologically or through manufacture so as to obtain information on their relative importance. In particular it will be interesting to modify interactions between components at different levels of structure and determine their criticality for fracture at the global level. Thus scale effects can be accommodated.

We eventually wish to use this approach to generate a variety of model materials using combinations of the controlling factors to achieve and optimise various mechanical properties such as stiffness, strength, toughness, etc. These new materials can be either immediately used as soft materials or as steps towards the design of stiffer composites, as happens when collagenous tissues are mineralised (e.g. turkey tendon, bone).

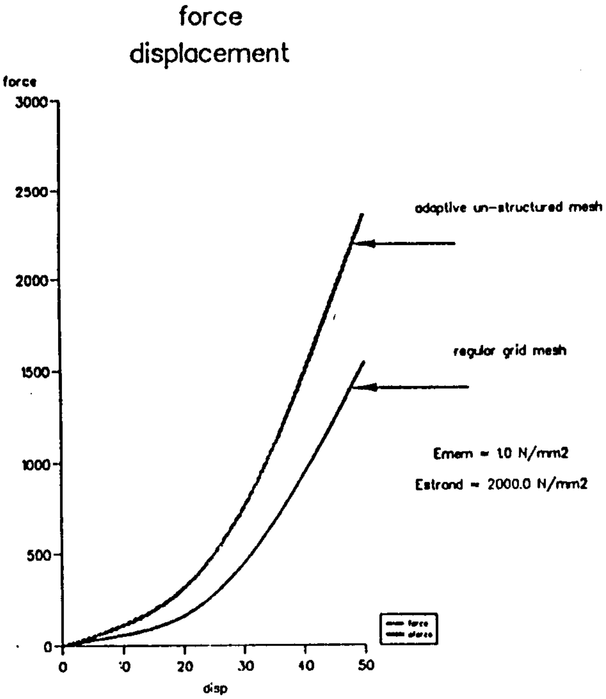


Fig. 5: Force-displacement curves derived from the data shown in figures 3 and 4.

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