THE MEASUREMENT AND MODELLING OF THE MECHANICAL PROPERTIES OF HUMAN SKIN IN VIVO—II. THE MODEL

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Abstract—The stress-strain relationship for human skin in vitro has a characteristic non linear shape even for low loads. Considerations are given on the basis of which a structural model has been selected, in which the mechanical properties of corrugated collagen fibrils are involved. It is found that such a model can describe the experimental stress-strain relationship surprisingly well with only three free parameters. These parameters are related to basic collagen fibril properties such as stiffness, diameter and waviness. The role of elastin is likely to be negligible for the purely elastic properties of human skin in vitro.

INTRODUCTION

The skin consists mainly of the dermal fibrous proteins collagen and elastin embedded in an amorphous ground substance. In relaxed skin the collagen fibres, which are arranged roughly parallel to the epidermis, have a wavy appearance (Finlay, 1969; Brown, 1973). Upon stretch the waveform disappears and the fibres tend to line up with the stress direction (Gibson and Kenedi, 1967; Finlay, 1969). The gradual straightening of the collagen fibres results in an increasing skin stiffness. After loading the fibres return to a wavy configuration.

In the preceding paper (part I, Manschot and Brakkee, 1986) it was described how the mechanical properties of human skin in vivo can be determined by the use of high uniaxial loads. Furthermore it was presented how to obtain a stress-strain relationship which is independent of the in vivo measuring configuration. In the same paper the viscoelasticity of skin was discussed. A procedure was given by which the experimental data could be split into time dependent and time independent data. These time independent properties are reflected in the purely elastic stress-strain relationship, which will be studied in the present paper. Such a relationship can be quantified by a mechanical model, which can be of a phenomenological or of a structural nature. Since, the latter offers the possibility to give a physical interpretation of the model parameters, the structural approach has been preferred.

In this paper considerations which have led to the choice of the specific model and the model's formulation will be presented. Furthermore a comparison between model and experimental data will be discussed in view of the morphology of the skin.

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For these reasons a model has been adopted which is based on the assumption that for unwinding each of the originally undulated collagen fibrils an increasing stress is needed (like a corrugated metal wire). A similar assumption has been made by Diamant et al. (1972), Ling and Chow (1977) and Comninou and Yannas (1976). In such a model, the waviness is an intrinsic property of the collagen fibril itself. Rigby et al. (1959) reported that, by gently teasing, rat tail tendon fibre bundles could be separated into parallel subunits, which showed the same waviness as the intact bundles. They further pointed out that the strain required to straighten out this waviness would be much larger than found experimentally if the pattern were helical. Also, micrograph studies on fibre bundles of tendon (Vidik, 1972; Diamant et al., 1972) did not provide any evidence for a helical arrangement. Since it is not likely that a sharp kinking exists in vivo (Vidik, 1980), modelling the tendon fibrils in a planar sinusoidal waveform seems to be the most appropriate.

When one deals with structures such as skin, in which the fibres are arranged in a multidirectional way, the orientation of the fibres with respect to the load axis may also be taken into account. However, this is not necessary in our experiments for two reasons. Firstly, most of the fibres in the skin of the human calf lie approximately in one direction, i.e. along the main axis of the leg (Manschot et al., 1982). Furthermore, the fibres orient upon stretch into the stress direction and remain in their new position after loading with high uniaxial loads (Craik and McNeil, 1966; Brown, 1973). Secondly, it can be shown (Manschot et al., 1982) that the contribution of fibres to the stress–strain relationship diminishes rapidly when the angle between fibre and load axis increases.

Therefore, the assumption has been made that after the preconditioning of the skin with a high uniaxial load the skin may be represented by a number of parallel aligned collagen fibres. Hence the straightening and the stretching of the skin fibres in such a situation is similar to that of the tendon fibres.

**Formulation**

The collagen fibrils in the skin are considered to behave like elastic springs with a periodic corrugation. The corrugation is given by a planar sinusoidal waveform and the material of the spring (i.e. collagen) is assumed to be linear elastic. Such a model has been described by Comninou and Yannas (1976), who did not however compare the model with experimental data. Although their model was developed for the description of tendon mechanics, it is essentially the same as presented in this paper and detailed derivation of the mathematical expressions can be found in their paper.

In the relaxed situation (schematically depicted in Fig. 1a) the waveform of the fibril is given by

\[ y = a \sin (bx) \]

The length of the fibril is considered to be long in comparison to the wavelength

\[ \lambda = \frac{2\pi}{b} \]

The maximal slope of the relaxed sinusoidal wave is

\[ \tan \theta = ab \]

Applying a force \( F \) in the \( x \)-direction at the ends of the fibril (with cross-section \( A_f \)) the wave will deform (Fig. 1b). For this situation the relationship between the stress \( F/A \) and the strain \( \varepsilon \) (relative increase in length along the \( x \)-axis) can be derived. In order to obtain a stress–strain relationship for the whole skin one has to consider that a force is applied to the whole cross-section \( A \) of the skin (stress \( \sigma = F/A \)). Consequently the collagen stiffness of the fibril \( (E_f) \) is related to the stiffness of the collagen of the skin \( E_c \) by

\[ E_c = \frac{N A_f}{A} \]

in which \( N \) represents the total number of fibrils. Hence, \( N A_f/A \) is a measure for the volume fraction of collagen in the skin. Defining the parameters

\[ \mu = \frac{a^2 b^2}{4} \]

and

\[ CF = A_f (b^2 I) \]

in which \( I \) is the moment of inertia of the cross-section \( A_f \), the stress–strain relationship of the skin is

\[ \varepsilon = \frac{(\mu + 1)}{1 + \mu \gamma^2} \left( \frac{E_c}{E_f} + 1 \right) \]

with

\[ \gamma = \frac{1}{CF \left( E_c (E_f) - 1 \right) + 1} \]
Since skin stiffness \( (E^*) \) can be defined as the derivative of the stress to the strain, one obtains from equation (2)

\[
E^* = \frac{E_c(1 + \mu \gamma^2)}{(1 + \mu)[1 + \mu \gamma^2(1 + 2\gamma Q)]}
\]

with

\[
Q = CF \left(\frac{\sigma}{E_c} + 1\right)\left(\frac{2\sigma}{E_c} + 1\right).
\]

For high stress values \( \gamma \) reduces to zero and the stress-strain relationship can be represented by a straight line (see equation 2) with a slope equal to \((1 + \mu)/(E_c)\). The intersection of this line with the strain axis is \( \mu \). Consequently \( \mu \) can be regarded as a strain and will therefore be expressed in per cent. It can be demonstrated that the parameter \( CF \) determines largely the ratio between the initial skin stiffness (for \( Q = 0 \)) and the ultimate stiffness.

The three parameter stress–strain relation (equation 2) is used to describe the mechanical properties of the human skin. The model parameters \( E_c, CF \) and \( \mu \) are estimated from the experimental stress–strain data using a nonlinear least square method based on the computer program BMDOTR (Dixon, 1974).

RESULTS

The model was applied to the results of strain experiments in 61 normal subjects of both sexes and various ages.

Typical results of the fitting procedure can be seen in Fig. 2 for the directions along and across the tibial axis. The open circles are the experimental data while the model is represented by the continuous line. The corresponding stress–stiffness curves have been plotted in Fig. 3. For some subjects the ultimate constant stiffness level is less pronounced. In that case the fitting is less accurate. However, in all cases the agreement between model and measured data is very close (correlation coefficients about 0.998). For the typical example of Fig. 2 the resulting parameters are

- along: \( E_c = 22 \text{ MNm}^{-2}, CF = 220, \mu = 8\% \);
- across: \( E_c = 6 \text{ MNm}^{-2}, CF = 100, \mu = 20\% \).

DISCUSSION

Since the model parameters can be interpreted in terms of collagen properties, these values can be compared to data concerning geometry and stiffness of the collagen fibril as they are known from literature. The parameter \( E_c \) determines largely the slope of the stress–strain curve for high stress values. The value for \( E_c \) (along) corresponds to ultimate stiffness data from uniaxial strain experiments of human skin \textit{in vitro} (Jansen and Rottier, 1958; Ridge and Wright, 1955;
Holzman et al., 1971). \( E_\alpha \) in the model represents the stiffness of the collagen fraction. Young's moduli for collagen from about 100 MN m\(^{-2} \) (Burton, 1968; Caro et al., 1978) to even 1000 MN m\(^{-2} \) (Torp et al., 1975) have been reported. Since the volume fraction of collagen in the skin is about 30\%, an ultimate stiffness of 30 MN m\(^{-2} \) or more could be expected. The difference between the observed values and the expected one indicates that only a part of all fibres are involved in the straining process in one direction. This may be due to the angular distribution of the fibres, which results also in a difference in stiffness between 'along' and 'across'. It may also originate from a non-uniform fibre geometry, by which only a part of all the fibres are straightened and stretched; e.g. Brown (1973) observed by microscope that in \textit{in vitro} experiments, the fibres in the mid dermal region are strained slightly in advance of those in the deeper dermal layers.

The parameter \( CF \) is a typical measure of the corrugation of the fibril and is, in contrast to \( E_\alpha \), independent of the total number of fibrils. If a circular cross-section of radius \( R \) is supposed for the fibril, then

\[
A_f = \pi R^2 \quad \text{and} \quad I = \frac{\pi R^4}{4}.
\]

Substitution in equation (1) gives

\[
CF = \left( \frac{\lambda_0}{\pi R} \right)^2.
\]

The wavelength of the fibril in the skin of the human calf amounts to about 3 \( \mu \text{m} \) (Gathercole and Keller, 1974). Combining this value with the observed parameter values one finds for the mean fibril diameter: along the tibial axis 120 nm and across 200 nm. Such mean diameters are reasonable considering the range of diameters in skin is from 50 nm to 400 nm (e.g. Oxlund, 1983).

The difference between the values along and across need not to be a result from a different (mean) fibril diameter. It is more likely that it results from a different wavelength in both directions (see further).

The parameter \( \mu \) indicates a strain value at which the mean corrugated fibril has become almost straight and is related to the maximal angle of the relaxed wave form (Fig. 1a).

From the observed values for \( \mu \), one then obtains for \( \theta_0 \)

- along: \( \theta_0 \approx 30^\circ \);
- across: \( \theta_0 \approx 40^\circ \).

Since \( CF \) is proportional to the square of wavelength \( \lambda_0 \), while \( \mu \) is inversely proportional to \( \lambda_0^2 \), their product \( CF \times \mu = (a/R)^2 \) is independent of \( \lambda_0 \). By a comparison of \( CF \) and \( \mu \) from 61 healthy subjects, no correlation between these parameters could be demonstrated. However, there was a significant correlation \((p < 0.005)\) between the products along and across. This indicates that at least the ratio \((a/R)\) is likely to be independent of the measuring direction. Assuming that such an independence is caused by both \( a \) and \( R \) independent of the direction, then the differences along and across in the parameters \( CF \) and \( \mu \) are due to a different wavelength.

In the model as presented the elastin fibres do not play a role, only the collagen fibres determine the purely elastic behaviour. From the close agreement between the model and the experimental data it follows that the contribution of the elastin fibres cannot be substantial. This conclusion is supported by the quantitative consideration on the possible involvement of the elastin fibres in human skin as given before. Moreover it was found that a significant correlation exists between the ultimate skin stiffness and the initial skin stiffness (Manschot, 1985). This also points to the conclusion that elastin is not relevant for the (initial) stiffness. The elastin fibres, however, may play a role in the viscoelastic process after loading. This has been confirmed in uniaxial strain experiments in rat skin \textit{in vitro} before and after enzymatical treatment with buffered elastase (Oxlund et al., in press).

REFERENCES


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