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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 vivo* 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 vivo*.

INTRODUCTION

The skin consists mainly of the dermal fibrous proteins collagen and elastin embedded in an amorphous groundsubstance. 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 *in*dependent 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 prefered.

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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THE MODEL

Considerations

The model to be considered will account for the purely elastic properties of skin, since the time independent stress–strain relationship will be studied. Hence, models with rate dependent features, such as fibres gliding through the ground substance or interfibrillar gliding, need not to be involved in these considerations.

Within a structural approach some authors depart from the assumption that the originally undulated collagen fibres do not transmit any force as long as they are not fully straightened (e.g. Lake and Armeniades, 1972; Decraemer *et al.*, 1980). In such a view the initial skin stiffness has to be explained by the stiffness of the elastin fibres or by the stiffness of some amount of collagen fibres. The latter implies that some of the collagen fibres have to be straight already at the onset of stretching. This is not likely since it would mean that, in the uniaxial strain experiments of the skin, these initially straight fibres become stretched for 20% or more, which is far beyond the rupture strain of collagen (a few per cent). On the other hand, the coefficients of elasticity found in the skin for minute deformations are much too high to be based solely on elastin. The values of such an initial stiffness for normal subjects are about 0.1–2 MNm^{-2} (Wijn, 1980; Jagtman, 1983). For elastin Young's modulus is about 0.3 MNm^{-2} (Burton, 1968; Caro *et al.*, 1978). Since skin consists only for a few per cent of elastin (Weinstein and Boucek, 1960), the modulus of elasticity would be at most 0.01 MNm^{-2} if only elastin was involved. In addition the elastin fibres probably have no meaning for the waviness of the collagen, since also such virtually elastin free tissues as the rat tail tendon bundle (Rigby *et al.*, 1959) and the free edge of the porcine heart valve leaflet (Broom, 1978) show such waviness, which is restored by relaxing the tissue.

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 (Viidik, 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* (Viidik, 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

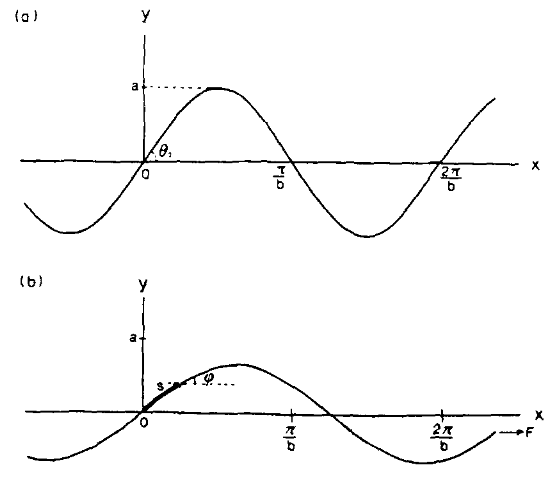

Fig. 1. Geometry of a model collagen fibril. (a) In a relaxed state. (b) In a deformed state.

comparison to the wavelength

$$\lambda\left(=\frac{2\pi}{b}\right).$$

The maximal slope of the relaxed sinusoidal wave is

$$tg\,\theta = ab.$$

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_f and the strain ε (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

$$E_c = E_f \frac{NA_f}{A},$$

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

$$\mu = \frac{a^2 b^2}{4} \tag{1}$$

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

$$\varepsilon = \frac{(\mu+1)\left(\dfrac{\sigma}{E_c}+1\right)}{1+\mu\gamma^2} - 1, \tag{2}$$

with

$$\gamma = \frac{1}{\left[CF\dfrac{\sigma}{E_c}\left(\dfrac{\sigma}{E_c}+1\right)\right]+1}.$$

Since skin stiffness (E^*) can be defined as the derivative of the stress to the strain, one obtains from equation (2)

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

with

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

For high stress values γ 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 μ. Consequently μ 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 $\sigma = 0$) 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 μ are estimated from the experimental stress–strain data using a nonlinear least square method based on the computerprogram BMDO7R (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: $E_c = 22\ \mathrm{MNm^{-2}}$, $CF = 220$, $\mu = 8\%$;
across: $E_c = 6\ \mathrm{MNm^{-2}}$, $CF = 100$, $\mu = 20\%$.

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 *in vitro* (Jansen and Rottier, 1958; Ridge and Wright, 1955;

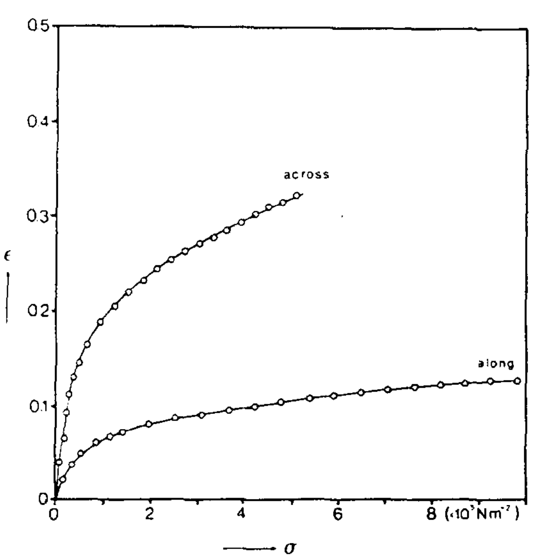

Fig. 2. Stress–strain relationships along and across the main axis of the leg. Open circles: experimental data, continuous line: model.

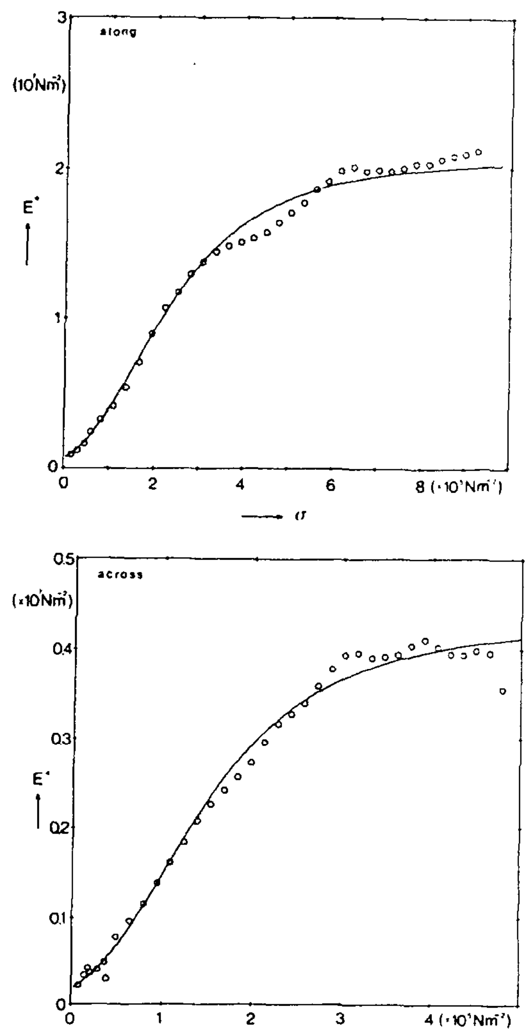

Fig. 3. Stress–stiffness relationships derived from the data of Fig. 2.

Holzman *et al.*, 1971). E_c in the model represents the stiffness of the collagen fraction. Young's moduli for collagen from about 100 MNm^{-2} (Burton, 1968; Caro *et al.*, 1978) to even 1000 MNm^{-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 MNm^{-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 *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_c, 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 μm (Gathercole and Keller, 1974). Combining this value with the observed parameter values one finds for the mean fibril diameter: along the tibal 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 μ 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 μ, one then obtains for θ_0

along: $\theta_0 \simeq 30°$;
across: $\theta_0 \simeq 40°$.

Since CF is proportional to the square of wavelength λ_0, while μ is inversely proportional to λ_0^2, their product $CF \times \mu = (a/R)^2$ is independent of λ_0. By a comparison of CF and μ 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 μ 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 *in vitro* before and after enzymatical treatment with buffered elastase (Oxlund *et al.*, in press).

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