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THE MEASUREMENT AND MODELLING OF THE MECHANICAL PROPERTIES OF HUMAN SKIN *IN VIVO*—I. THE MEASUREMENT

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Abstract—The mechanical properties of human skin *in vivo* are studied by means of uniaxial strain measurements. In order to obtain a stress–strain relationship which is independent of the *in vivo* measuring configuration, values for the effective width and effective length of the loaded skin strip have to be known. By variation of tab width and tab distance in a few series of experiments on the same subject, these effective values are found.

In order to obtain a time *in*dependent stress–strain relationship a correction procedure is introduced. In this procedure the time dependent (viscoelastic) effects are described and subtracted from the total response.

INTRODUCTION

In a previous investigation Wijn (1980) studied the mechanical properties of human skin *in vivo* for small deformations (strains of a few per cent and loads of maximal 0.05 MNm^{-2}). The present work deals with uniaxial strain measurements of the skin of the human calf for large deformations (up to 50%) using high loads (up to 1 MNm^{-2}). In order to obtain mechanical parameters of the skin which are independent of the size of the loaded skin strip, the force and the elongation have to be converted into stress (force divided by cross-sectional area) and strain (relative increase in length). For this purpose the dimensions of the skin strip have to be known. Since *in vivo* the skin can only be strained by pulling apart two tabs which are attached at the skin surface, the effective length of the strip may be different from the distance between the tabs. In addition the effective width of the skin strip may be larger than the width of the tabs if the adjoining skin is also deformed. For these reasons attention will be paid to the quantification of the effective dimensions of the strained skin strip *in vivo*. Furthermore a method for the signal analysis will be presented enabling the calculation of the purely elastic, time independent, stress–strain relationship from the total response.

Since the skin of the calf exhibits, besides non-linearity and viscoelasticity, also anisotropy the stress–strain relationship will be determined in two directions: along and across the tibial axis. These directions coincide roughly with the directions of maximal and minimal skin stiffness in this area (Manschot *et al.*, 1982).

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METHODS

Apparatus

In our uniaxial strain experiments two square tabs (10 × 10 mm) are attached to the skin of the human calf by means of a cyanoacrylate adhesive. The distance between the tabs is 5 mm. One of the tabs is fixed to the frame of the apparatus (Fig. 1) and the other one is connected to a permanent magnet which can move freely along the axis of a cylindrical coil, parallel to the skin surface. The tabs are pulled apart by electromagnetic forces generated with a programmable current source in connection to the coil. These forces are independent (< 1%) of the position of the tab over a range of 12 mm. The resulting displacements of the tab are detected by an inductive displacement transducer. Forces and displacements are sampled on line using a DEC PDP 11/34 computer.

The uniaxial strain apparatus is mounted on a couch in such a way that the tabs can easily and reproducibly be attached to the calf of a subject lying on the couch. The adhesion of the tabs to the skin is improved by cleaning the skin beforehand with a tissue moistened with petroleum ether. During the experiments the tabs

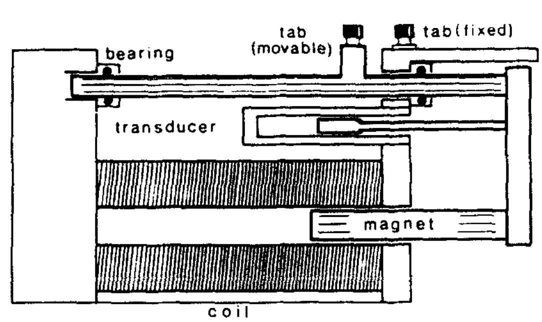

Fig. 1. A schematic drawing of the uniaxial strain apparatus.

are lightly indenting the skin (about 1/2 mm). Even for the highest loads the measurements give no discomfort (or pain) to the subject.

Experiment

Four identical sawtooth shaped loads with a maximal amplitude of 12 N are used. The load duration is 10 s and the interval time is 20 s. In order to be able to separate the purely elastic and the viscoelastic effects in the sawtooth response, the loading is completed with six small pulse shaped loads of increasing amplitudes (up to 1 N).

The responses of the skin to the second and subsequent sawtooth shaped loads are very similar but differ from the first one. This first response, reflecting the preconditioning phase (Fung, 1972) is left out of consideration.

In vivo *skin strip dimensions*

The local skin thickness is measured by an ultrasound technique (Alexander and Miller, 1979) using a polyvinylidene fluoride (PVDF) transducer as described by Payne and Quilliam (1983). Skin thickness is calculated from the time between the arrival of two ultrasound echoes: one originating from the skin surface and one from the dermal/sub-dermal junction. The accuracy in skin thickness determination with this technique is about 0.1 mm. A typical value for skin thickness is 1.2 mm.

In order to determine the width of the skin strip, it has been assumed that any involvement of the tissue bordering one side of the skin region just between the tabs, is not influenced by the tissue on the other side of that region. Consequently, an additional width is independent of the width of the tabs itself. In series of experiments with tabs of various widths, but at the same initial distance, a linear relationship exists between force F (at a constant elongation) and width b (see also Manschot, 1985). In these experiments b is varied from 5 to 20 mm. With every width the related force values amount to about 10 N. The intersection of the regression line with the b-axis gives the additional width. This was done in both directions (along and across the main axis of the leg) and with various initial distances and elongations. In none of the experiments a significant width could be demonstrated, not even for small deformations. So the effective width of the skin strip is equal to the width of the tabs. Using the values for thickness and width, the applied force can be converted into a stress.

The effective length of the strip at any moment is the sum of the effective initial length (just before the onset of the load) and the effective deformation Δl_{eff}. In experiments using only small deformations it could be demonstrated that using a cyanoacrylate adhesive the initial length equals the initial tab distance. This means that the effective point of application of the forces is right at the edge of the tabs.

The increase in length of the skin strip all over its thickness may differ from the tab displacement Δl_{tab}. In fact, the increase in length in the deeper layers of the dermis will be less as compared to that at the surface (Fig. 2). This effect can be described by a (mean) residual deformation Δl_{res}, i.e.

$$\Delta l_{eff} = \Delta l_{tab} - \Delta l_{res}$$

in which Δl_{eff} represents the effective (mean) increase in length. It seems reasonable to assume that the lag of the lower layers with respect to the tab displacement will depend on the force exerted on the connection between the different layers and not on the extent of the tab displacement nor on the initial distance of the tabs. Proceeding from this assumption a number of experiments were carried out using various initial tab distances (3–20 mm) but still the same stress value. Thus, a value for Δl_{res} belonging to this stress value can be found by plotting the observed tab displacements as a function of the initial tab distance. From a comparison of Δl_{res} for different stresses it was found that Δl_{res} depends linearly on the stress (Fig. 3). Furthermore it was found that Δl_{res} was independent of the width of the tabs but did depend on skin thickness. An appropriate relationship is

$$\Delta l_{res} = 2.3 \times 10^{-7} \times d \times \sigma \text{ (mm)},$$

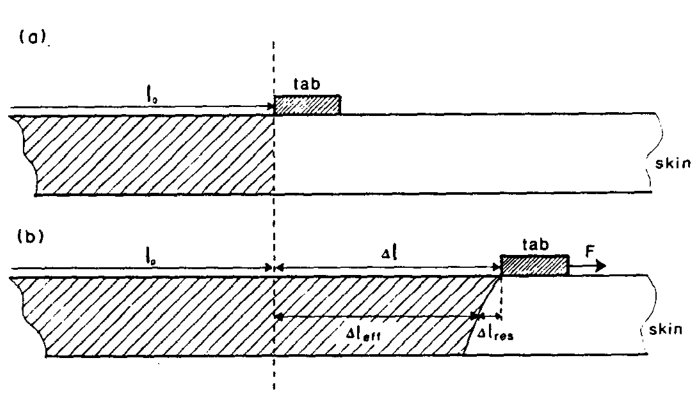

Fig. 2. A representation of the cross-section of the skin in an uniaxial strain experiment. (a) Relaxed situation. (b) Strained situation.

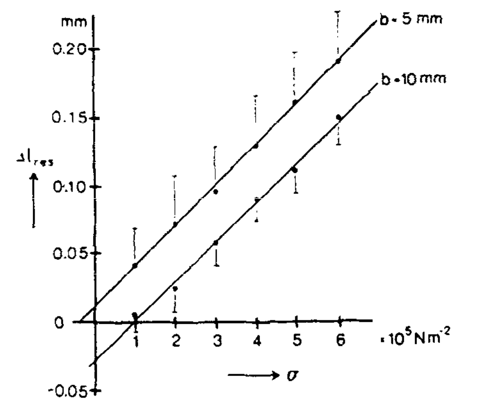

Fig. 3. Δl_{res} vs σ plots for two values of the width of the tabs.

with skin thickness d expressed in mm and the stress σ in Nm^{-2}. This correction of Δl_{tab} is used for the calculation of the effective strain.

The time dependent effects

Human skin has viscoelastic properties. This can be seen clearly in the unloading phase of the response on a sawtooth shaped load (Fig. 4, load offset at $t = 10$ s). Three regions can be distinguished: a purely elastic deformation ε_a, a viscoelastic deformation ε_b and a permanent deformation ε_c. (In this context the term 'permanent' means long as compared to the interval time of 20 s.) The same three processes are present during loading. Hence, the purely elastic strain $\varepsilon_a(\sigma)$ as a function of the stress, which is proportional to time, can be obtained by correcting the total strain at every instant during loading for the time dependent effects ε_b and ε_c.

Correction for ε_c. The well known governing differential equation for a purely viscous process is (Flügge, 1975)

$$\sigma(t) = \eta_s \frac{d\varepsilon_c(t)}{dt} \tag{1}$$

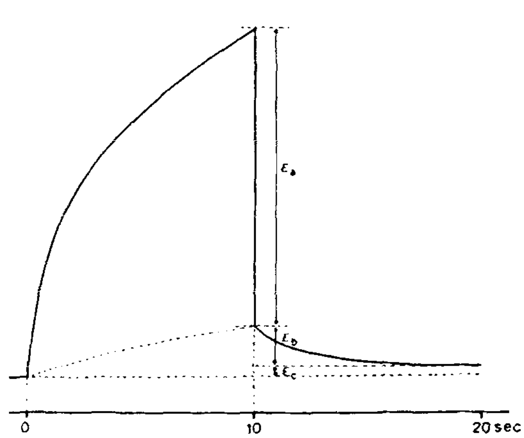

Fig. 4. Schematic drawing of response of the skin to a sawtooth shaped load with a duration time of 10 s.

in which $\sigma(t)$ represents the stress and η_s a coefficient of viscosity. In a preliminary study it was found that within the time period of the experiment and within the range of the stresses as used, η_s is independent of time and stress, which is in accordance with the findings of Vlasblom (1967). Consequently the solution of (1) for a sawtooth shaped load ($\sigma(t) = \hat{\sigma}t/T$ for $0 < t < T$ and $\sigma(t) = 0$ otherwise, with load duration T and $\hat{\sigma}$ constant) is

$$\varepsilon_c(t) = \frac{1}{2}\frac{\hat{\sigma}}{T\eta_s}t^2 \quad \text{for} \quad 0 < t < T. \tag{2}$$

The value for η_s can be determined from the observed permanent deformation at the end of the unloading phase, since for $t \geqslant T$

$$\varepsilon_c(t) = \varepsilon_c = \frac{1}{2}\frac{\hat{\sigma}}{\eta_s}T.$$

Correction for ε_b. In experiments with pulse shaped loads (constant stress during loading) it was found that the delayed elastic deformation both during loading and unloading can be described by a single exponential time function with time constant τ and amplitude ε_b. From a comparison of the pulse responses on different stress magnitudes it was found that τ was constant (about 3.5 s) over the whole range of stresses. From the same experiments it was found that ε_b is nonlinearly related to the stress. This nonlinearity can be characterized by a logarithmic relationship (Wijn, 1980; Manschot, 1985)

$$\varepsilon_b = 1/k \ln\left(1 + \frac{k\sigma}{E_0}\right) \tag{3}$$

a relationship which is valid for a number of soft tissues (Fung, 1967).

The governing differential equation for the viscoelastic process is given by

$$g(\sigma) = \varepsilon_b(t) + \tau\frac{d\varepsilon_b(t)}{dt} \tag{4}$$

in which τ is a constant and $g(\sigma)$ represents the amplitude of the deformation for a stress $\sigma(t)$. For a sawtooth shaped load this function $g(\sigma)$ is equal to ε_b of equation (3) in which σ is replaced by $\hat{\sigma}t/T$. Consequently $g(\sigma)$ is also a function of time.

The solution of equation (4) for a sawtooth shaped load is equal to

$$\varepsilon_b(t) = \frac{1}{k}\ln(1 + Bt)\left(1 - \exp\left(-\frac{1+Bt}{B\tau}\right)\right) - \left[\frac{1}{k}\exp\left(-\frac{1+Bt}{B\tau}\right)\sum_{n=1}^{\infty}\left(\frac{1}{B\tau}\right)^n\frac{(1+Bt)^n - 1}{n.n!}\right]$$

with $B = \dfrac{k}{E_0}\dfrac{\hat{\sigma}}{T}$.

The parameters E_0, k and τ can be estimated [by using equation (3)] from the experimental data in which the responses of the six small pulse shaped loads have to be included. In practice a summation over ten terms in equation (5) is sufficient for the correction.

The purely elastic stress–strain relationship is obtained by subtracting $\varepsilon_c(t)$ and $\varepsilon_b(t)$ (with respect to equations 2 and 5) from the total response of the skin at every instant during loading.

RESULTS

Two thus corrected stress–strain relationships of the skin of the calf *in vivo* are shown in Fig. 5, showing respectively the behaviour of the skin along and across the tibial axis. Defining the stiffness of the skin E^* by $E^* = d\sigma/d\varepsilon$ the corresponding stress–stiffness curves can be calculated (Fig. 6). The curves shown in Figs 5 and 6 are all obtained from effective deformation data.

It can be noticed that skin clearly exhibits nonlinear and anisotropic properties. For both directions skin stiffness increases at low stresses strongly and almost linearly with stress [in correspondence with Fung (1967, 1972)], but levels off toward an ultimate value for high stresses.

DISCUSSION

In order to obtain a purely elastic, configuration independent stress–strain relationship for the skin *in vivo* some corrections on the experimental and configurational data have to be carried out.

More or less surprising was the finding that the involvement of the skin adjoining the skin strip just between the tabs could be neglected. In other words: the effective width of the piece of skin is equal to the width of the tabs. Since from a mechanical point of view the skin mainly consists of collagen and elastin fibres embedded in an amorphous groundsubstance, this result indicates that after the application of a high preconditioning load only the fibres running approximately parallel to the direction of stretch are

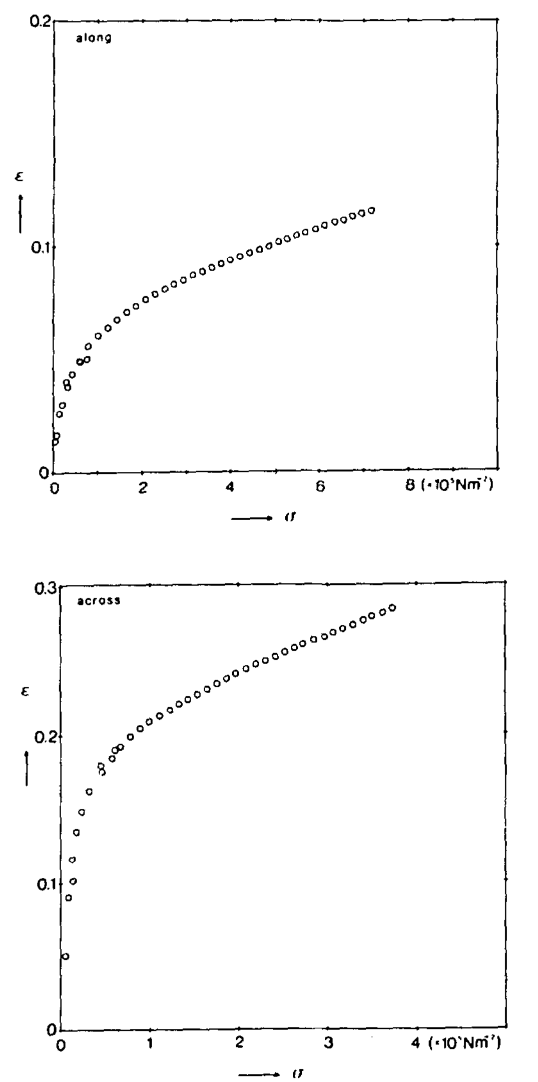

Fig. 5. Purely elastic stress–strain relationships along and across the tibial axis.

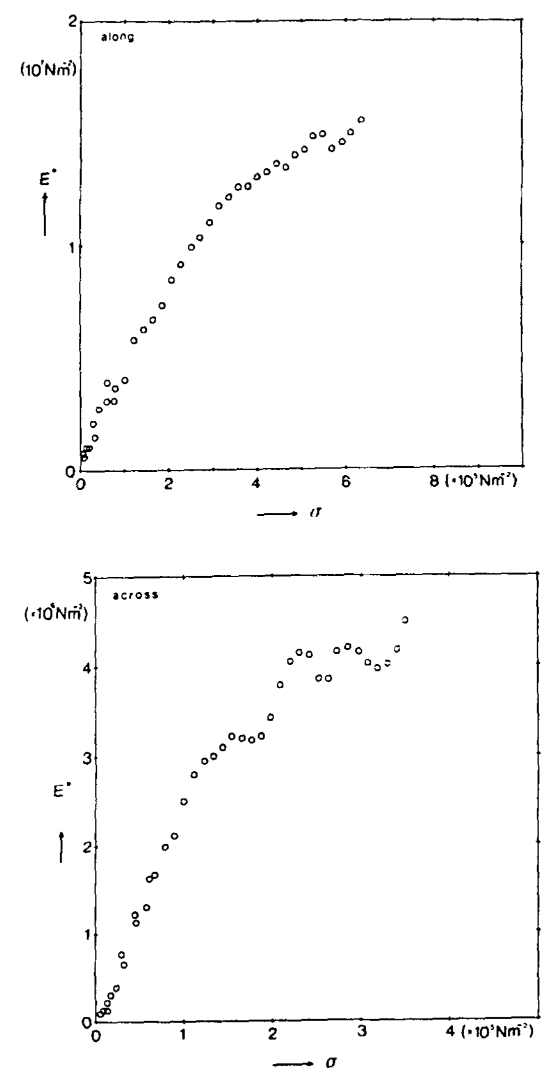

Fig. 6. Purely elastic stress-stiffness relationships along and across the tibial axis.

involved. Without such a high preload the effect of the outside area cannot be neglected (Wijn, 1980). This result can be understood within the view of fibres shifting through the groundsubstance into the direction of stretch as has been observed by microscope by a number of authors (Finlay, 1969; Brown, 1973). Since no effect of the outside area could be detected even some minutes after preloading, it can be concluded that for such periods the shifting is not reversed. This conclusion is supported by the microscope findings of Craik and McNeil (1965).

The correction for the lag in deformation of the lower skin layers is necessary, especially in the case of high loads. This becomes evident by comparing stiffness curves obtained in experiments on the same subject but with different initial tab distances. After the correction the highly different curves became almost the same, as could be expected if indeed pure skin properties are measured. It will be clear that such a correction is strongly needed e.g. when comparing *in vivo* and *in vitro* data, since for the latter mostly clamps are used, by which the skin strip is equally strained over all its thickness.

For large stress values the stress–stiffness curves tend to reach a plateau; the stress–strain relationships become almost linear. Since it is generally accepted that for high stresses the mechanical properties of skin mainly reflect collagen properties, our result is in agreement with the findings of Abrahams (1967) and Viidik (1979) that the load–elongation relationship of fully aligned collagen fibres approaches linearity. Also the results of Danielsen (1982), who observed similar stress–stiffness curves for membranes composed solely of reconstituted collagen fibrils, leads to the conclusion that the method as described enables the determination of the properties of the collagen fibres of the skin *in vivo*.

The results correspond to data as observed in uniaxial strain experiments of human skin *in vitro* (Jansen and Rottier, 1958; Ridge and Wright, 1966; Holzman *et al.*, 1971). Also the stress needed to reach the linear region of the stress–strain curve is comparable to that given by Brown (1973): about $3 \times 10^5\ \mathrm{Nm}^{-2}$. A model description and a quantitative discussion of these results is given in a second paper.

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