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Measuring the viscoelastic properties of tissue using changes in its capacitance: A feasibility study

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Abstract. Measurement of the electrical and viscoelastic properties of tissue is important in determining its physiological state. We present here a proof of principle study of a new technique that yields both sets of properties for soft tissue in vivo, with minimal discomfort. The variation of skin capacitance with time after the application of pressurized electrodes is used to determine the viscoelastic properties of the tissue. The mechanical compression of the tissue, measured by the capacitance increase with time, is related to the mechanical strain. Knowledge of the stress applied by the electrodes then yields the viscoelastic properties of the tissue corresponding to the standard, linear four-element viscoelastic model. Results are presented for three different subjects to show that the method is sensitive to individual variability. The series elasticity, the viscoelastic (parallel) elasticity and viscosity, and the series viscosity are determined for six applied stresses in the MPa range. Significant non-linearity is observed with the viscoelastic parameters increasing with the applied stress. This new method provides values for the viscoelastic properties of the skin comparable to those reported in the literature and could be of significant value for clinical diagnostics and for surgical modeling.

Keywords: skin, mechanical properties, creep test, impedance measurements, numerical model

Merjenje viskoelastičnih lastnosti biološkega tkiva

Z merjenjem električnih in viskoelastičnih lastnosti biološkega tkiva lahko določamo njegovo fiziološko stanje. V prispevku opisujemo metodo za neinvazivno in vivo merjenje električnih in mehanskih lastnosti tkiva. Metoda vključuje pritisk dveh ali več elektrod na kožo s konstantnim pritiskom in sočasno merjenje sprememb električne impedance s časom. Le-te sovpadajo z mehanskimi spremembami, ki jih povzroča pritisk elektrod na kožo, zato lahko impedančne meritve uporabljamo tudi za določanje mehanskih lastnosti kože. Viskoelastične lastnosti tkiva so predstavljene s standardnim linearnim modelom ki vključuje 4 elemente s katerimi opisujemo različne elastične in viskozne mehanske odzive tkiva. Z opisano metodo izmerjene vrednosti viskoelastičnih lastnosti kože so primerljive s tistimi, ki jih najdemo v literaturi. Potencialna uporaba metode je klinična diagnostika in modeliranje na področjih tkivne medicine.

1 INTRODUCTION

Measurement of the viscoelastic and electrical properties of tissues is important for their characterization. Viscoelastic measurements can be used to identify pathological conditions in tissues as well as to provide tissue parameters for surgical simulation, among other uses [1]. Similarly, electrical measurements can also be used for therapeutic purposes [2]. We propose here a method that provides both the electrical and the mechanical properties of tissue, in vivo or excised, with one set of measurements. Moreover, because the penetration of electric field is frequency-dependent, application of this new technique allows the characterization of tissue viscoelastic properties with depth by using several frequencies. We demonstrate the feasibility of this method by performing measurements on three subjects to show that it is capable of detecting individual variations.

We chose to use skin as the tissue for the validation of this method because its complex structure makes modeling a challenge. Skin is a very inhomogeneous, stratified tissue with several major layers [3]:

a) Subcutaneous tissue: Strictly speaking, this layer is not a part of skin tissue, but nevertheless contributes to its mechanical properties. It consists predominantly of loose connective tissue and adipose cells, acting as a shock absorber and heat insulator.

b) Dermis: This middle layer is the thickest (1-5 mm). It is made mainly of collagen and elastin fibers that are embedded in a ground substance that is a complex, extracellular material containing water, electrolytes and glycoproteins. The dermis is the main mechanical

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component of skin, responsible for its structural integrity, elasticity and resilience.
c) Epidermis: Its total thickness is usually about 30-80 μm. It consists primarily of keratinocytes in progressive stages of differentiation from deeper to more superficial layers. Mechanically it is quite rigid but often overlooked as an important contributing factor in the overall mechanical response of skin.
d) Stratum corneum: A dead outer layer about 15 μm thick (except on soles and palms: up to 1 mm) is made of keratinized cells. It is a hard material; its mechanical properties depend strongly on hydration and temperature.

The elastic properties of skin are determined primarily by the collagen and elastic fibers whereas the viscous properties are determined primarily by the tissue water and ground substance in the dermis [4].

The electrical properties of skin and other tissues are described in [2]. A thorough documentation of tissue electrical properties, their measurement and their modeling can be found in a three-part review [5]–[7]. Briefly, the electrical impedance of the skin is composed of two parts: the resistance that determines energy dissipation and the capacitance that determines energy storage. The resistance is associated with ionic conduction channels and is thus affected by skin hydration and water transport whereas the capacitance is determined by lipid or protein rich domains and is less sensitive to hydration and water transport [8], [9]. The stratum corneum has the highest resistivity of the skin layers. The rest of the epidermis, the dermis and the subcutaneous tissue all have much lower resistivities [10], [11]. In contrast, as will be shown below, the contribution to the capacitance of the skin layers, as determined by their permittivities and thicknesses, is lower for the stratum corneum than for the underlying layers. The contributions of different skin layers towards skin impedance differ depending on the frequency used. Studies show that for frequencies less than 10 kHz the share of stratum corneum in the total impedance of skin is around 50%, but at 100 kHz drops to around 10% [12].

The mechanical properties of skin have been measured by a variety of methods such as suction, torsion and traction [13]. We use indentation as it is less invasive than other techniques and produces less discomfort for in vivo measurements. Indentation techniques in general are commonly used for clinical purposes, such as prostate examinations [14], and for modeling surgical procedures [15]. Some recent studies have used Atomic Force Microscopes (AFM) to produce nanoindentations to measure excised skin viscoelastic properties on the microscale [16]–[18]. Because our proposed method is capable of making measurements both in vivo and on excised tissue, we compare our results to those obtained by indentation of human skin in vivo [13], [19]–[23], as those values are potentially more relevant clinically and for surgical modeling.

An early indentation study [19] used a weighted metal rod sliding in a vertical tube to produce indentation and a micrometer to measure it. More recently, several more precise types of indentation methods have been used to measure skin's viscoelasticity in vivo. In stress relaxation an indentation of fixed amount is suddenly applied and then held for a brief duration while the applied force necessary to maintain the indentation is measured as a function of time [20], [21]. An instrument has been developed that simultaneously measures the variation of the applied force and the resulting indentation [13], [22]. In the creep method a constant force is applied and the resulting change in penetration depth with time is measured [21], [23]. The technique proposed in this paper uses the latter, creep method with the capacitance change yielding the tissue's strain.

Electrical impedance measurements involve pressing two or more electrodes against the skin and measuring the variation of the electrical impedance, or related parameters, as a function of frequency. While conducting measurements of the inter-electrode resistance and capacitance, we noticed that the results varied systematically with time – the capacitance steadily increasing and the resistance steadily decreasing. In particular, the capacitance response with time mirrored the creep test curve. These capacitance changes coincide with the indentation of the surface and could be used to determine the mechanical properties of the skin while measuring its electrical properties. Further, as the sampling depth during impedance measurements on skin depends strongly on the frequency used [2], [12], such measurements could be used to assess the viscoelastic properties of different skin layers.

This paper describes in detail how the variation of capacitance with time reflects the compression of the skin described by the standard, linear four-element mechanical model. We validate the results by comparing them to those obtained by in vivo indentation methods and confirm the nonlinear variation of the viscoelastic properties with the applied stress. This paper demonstrates the feasibility of this methodology and illustrates the principles of the mechanical response by measurements made on three subjects to show that the method is sensitive to individual variability. Comparison of the mechanical properties for a broader population to determine gender and age effects, however, requires a much larger, future study.

2 MATERIALS AND METHODS

2.1 Electrical measurements

The tissue beneath each electrode is modeled as a parallel combination of a resistor, *R'*, and a capacitor, *C'*, as shown in Figure 1. The total resistance, R_p, is then *2R'* and the capacitance, C_p, is *C'/2*. Resistance *R'*, in turn, can be regarded as a series combination of the

resistance of the stratum corneum and the resistance of the epidermis+dermis. As the former is much greater than the latter, R_p is determined by the stratum corneum, except at frequencies in the high kHz range.

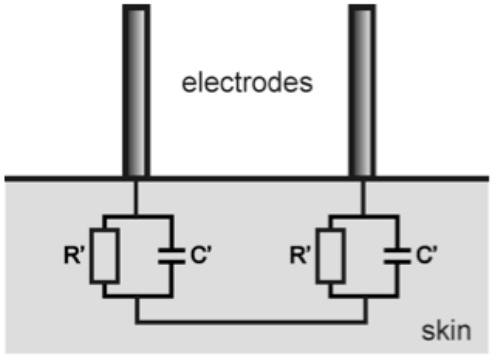

Figure 1: Electrical model of the skin beneath each electrode

The capacitance is more complicated. Assume for simplicity a parallel-plate system with relative permittivity κ, effective electrode area A and effective electrode separation d. In that case the capacitance, C, is given by

$$C = \kappa\varepsilon_0 A/d \quad (1)$$

where ε_0 is the permittivity of free space. Let index 1 denote the stratum corneum and 2 the viable skin. Then

$$1/C' = d_1/\kappa_1\varepsilon_0 A + d_2/\kappa_2\varepsilon_0 A \quad (2)$$

Typical values for d_1 and d_2 are about 10^{-5} m and 10^{-3} m, respectively [3]. κ_1 and κ_2 vary with frequency. At 1 kHz $\kappa_1 \sim 10^3$ and $\kappa_2 \sim 10^5$ while at 1 MHz $\kappa_1 \sim 200$ and $\kappa_2 \sim 3\text{x}10^3$ [24]. For these values at 1 kHz the contributions to C_p by the stratum corneum and the underlying tissues are comparable while at 1 MHz the underlying tissues dominate. Deeper penetration of signal at higher frequencies into skin has also been shown numerically [12].

The stratum corneum thus dominates resistance measurements, but is relatively less important for capacitance measurements. The electrical impedance of the skin is known to depend on a wide variety of factors, but at low frequencies this dependence is far more important for the resistance than for the capacitance [10], [11]. Moreover, as noted above [8], [9], the capacitance is less sensitive to hydration and water transport than the resistance. Unlike the capacitance, our measured variation of resistance with time was erratic below a few kHz. At frequencies well above a few kHz the results of a resistance analysis were comparable. Using capacitance as the measurement parameter minimizes the effects of changes in surface properties due to sweating, etc. and water transport in the electrical measurements. For this reason and because the capacitance values were more sensitive to the underlying tissues we chose to concentrate on the variation of capacitance with time.

2.2 Electromechanical analysis

The mechanical properties of skin also show a layer-dependence. When skin is subjected to sustained stress, its mechanical response can be divided into three phases: i) the immediate, purely elastic phase, ii) the viscoelastic phase of variable creep, and iii) the entirely viscous phase of constant creep. The typical variation of strain with time produced by the application of a constant mechanical stress is depicted in Figure 2(a). This kind of temporal response can be modeled by the standard, linear four-element mechanical model shown in Figure 2(b).

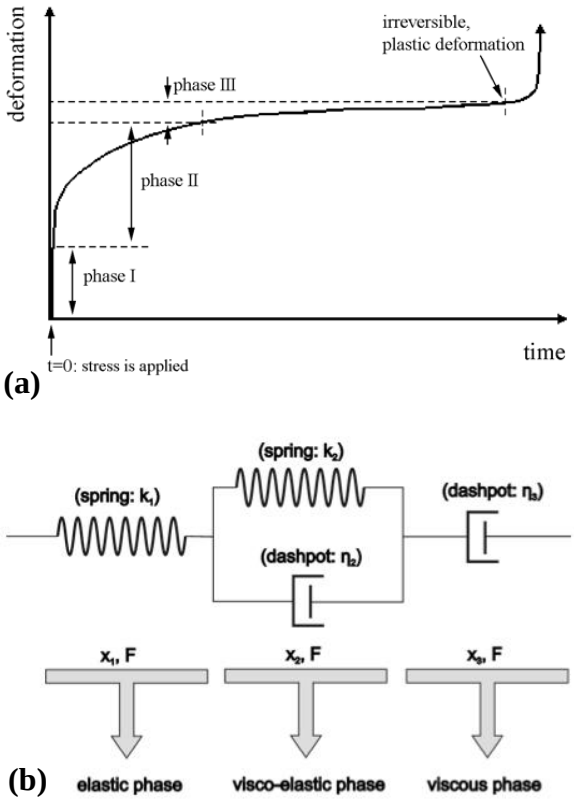

Figure 2: (a). Skin deformation versus time for constant applied stress. Phase I: immediate, elastic deformation, phase II: viscoelastic deformation, phase III: viscous deformation; (b). The analytical model used to describe the three phases of skin's mechanical response

Agache and Varchon [3] refer to a similar four-element model for the analysis of creep experiments on skin. Jachowicz et al. [21] ignored the series viscosity element and applied the three-element model to creep experiments on skin. Boyer et al. [22] used just the parallel, two-element viscoelastic model to analyze their results. In our case it was important to use the complete four-element model because the duration of our

experiments was sufficiently long for the series viscosity term to appear.

Analytically, the variation of strain with time, $\varepsilon(t)$, for this system is given by [25]:

$$\varepsilon(t) = \sigma/E_1 + \sigma(1-e^{(-t/\tau)})/E_2 + \sigma t/\eta_3 \quad (3)$$

where σ is the applied stress and $\tau=\eta_2/E_2$ is the viscoelastic relaxation time. The E are elastic moduli and the η are viscosities. The first term represents the initial, purely elastic response. The second is the subsequent viscoelastic compression. The last term is a late, purely viscous response.

As shown in Figure 3, the variation of measured capacitance versus time shows a pattern identical to that of the compression shown in panel (a) of Figure 2.

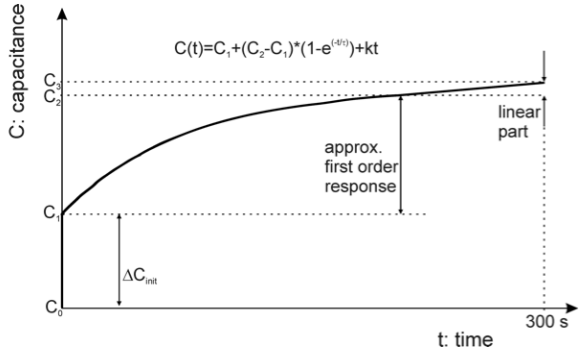

Figure 3: Computational model used to analyze the variation of skin capacitance with time

According to Equation 1 a steady increase in capacitance can be regarded as a decrease in d; that is, a compression of the tissue. Let d_0 be the original value of the effective separation and d_f the final value. Using Equation 2 we find that the mechanical strain ε is

$$\varepsilon = (d_0 - d_f)/d_0 = (C_f - C_0)/C_f \quad (4)$$

Thus the strain depends only on the ratio of measured capacitances and is independent of the actual κ and A values.

The initial, purely elastic strain, $\varepsilon_1=\sigma/E_1$ corresponds to $\Delta C_{init} = C_1 - C_0$ or, using Equation 4

$$E_1 = C_1\sigma/(C_1 - C_0) \quad (5)$$

The initial elastic strain, ε_1, plus the subsequent viscoelastic strain, ε_2, corresponds to C_2 - C_0 or

$$E_2 = \sigma/\varepsilon_2 \quad (6)$$

where, again using Equation 4:

$$\varepsilon_2 = (C_2 - C_0)/C_2 - \varepsilon_1 \quad (7)$$

The viscosity is then

$$\eta_2 = \tau E_2 \quad (8)$$

To determine the series viscous response term, η_3, we use the capacitance value, C_3, at a particular time, T = 300 s.

$$\eta_3 = \sigma T/\varepsilon_3 \quad (9)$$

where

$$\varepsilon_3 = (C_3 - C_0)/C_3 - \varepsilon_2 - \varepsilon_1 \quad (10)$$

2.3 Experimental methods

Impedance measurements on skin were performed by means of precision LCR meter HP4284A with a 20 Hz to 1 MHz measurement frequency range, controlled by Labview (National Instruments, Austin TX) program. Spring-loaded pin electrodes F773 (Feinmetall GmbH, Herrenberg, DE) doubling as mechanical probes were used to measure impedance changes during the mechanical deformation of skin. The pins were circular cylinders with flat ends. As noted by Boyer et al. [22], this shape minimizes adhesion to the skin and provides a constant contact radius. The spring loading could be adjusted to provide a known contact force. With known applied force and contact area the applied stress is easily estimated.

Two pin electrodes/probes were fitted in a circular housing covered with a flat surface, with a small opening for the pins. The support of the housing allowed for a controlled placement of the pins on the skin surface while the small aperture provided mechanical constraint to limit mechanical deformations to a small area between and around the electrodes/probes. The measurement setup and the electrodes with the housing are depicted in Figure 4.

Figure 4: The measurement setup.

The measurements were performed in a controlled environment; the temperature was maintained at 25 °C ±1 °C, with relative humidity at 50 %±5 %. In order to demonstrate that the method was sensitive to individual variability we performed the measurements on three subjects, two males and a female, from different age groups: 23-year old male (Y), 37-year old female (F) and 69-year old male (S). The subjects were left to acclimatize for at least 30 minutes before the experiment. Further, as different tissue depths are sampled at different frequencies, measurements were done at two frequencies: 400 Hz and 1 MHz, on the volar forearm. The study has been approved by the Slovenian Medical Ethics Committee (application number 104/07/13).

Different pins were used for the measurements, with different diameters (1.4 and 3 mm) and spring forces, which provided us with six different stresses (0.175 MPa, 0.238 MPa, 0.308 MPa, 0.50 MPa, 0.806 MPa and 1.09 MPa) to cause mechanical compression of the skin. The interelectrode distance (between pin centers) was 5 mm. Prior to compression, the initial capacitance was measured with the pins touching the skin to give C_0. The pins were then pressed perpendicularly against the skin for five minutes to cause mechanical deformation during which time impedance was recorded. Approximately two seconds were required for each reading of R_p and C_p at two frequencies: 400 Hz and 1 MHz.

We temporarily ignore the first few data points that correspond to the initial elastic compression and begin the fit at a time when the viscoelastic stages appear to begin. That time is very short (several seconds) compared to the viscoelastic time constant so little error is introduced. The subsequent variation of measured capacitance with time is modeled using the expression, analogous to Equation 3:

$$C(t) = C_1 + (C_2 - C_1)\cdot(1 - \mathrm{e}^{(-t/\tau)}) + kt \qquad (11)$$

where k is a constant. Since C_1, the capacitance at the start of the viscoelastic compression, is determined by the fit, the actual duration of the initial rise is not important. The fit is performed with a non-linear, least-squares fitting procedure [26] implemented using a VisualBasic routine in Excel (Microsoft, Redmond WA). It should be noted that in many cases it is not necessary to use the parameter k as the capacitance-time curve becomes constant, indicating that the series viscoelastic term is not important in such cases. Three parameters then suffice to fit up to 140 measured values very well. With C_1 determined from the fit, we can then use $C_1 - C_0$ to evaluate E_1 using Equation 5.

3 RESULTS

The focus of our measurements is to establish the relation between the viscoelastic properties of skin tissue and its electrical impedance. We begin by showing the change in inter-electrode capacitance upon compression of the electrodes. Figure 5(a) illustrates the 400 Hz fits for capacitance-time measurements made on the forearms of Y, F and S. The applied stress was 0.5 MPa. In each case the four parameters C_1, C_2, τ and k provide an excellent fit for up to 140 data points. The greatest difference (22%) between the measured and calculated values occurs at time t = 0 s. Because the initial elastic compression may have been completed in between the data points, the viscoelastic starting time is offset slightly. The average difference for the remaining points in the three fits is at most 1.4%. There is no clear dependence of the capacitance at 400 Hz between subjects, but the capacitance for the female subject is lower than those of the two male subjects.

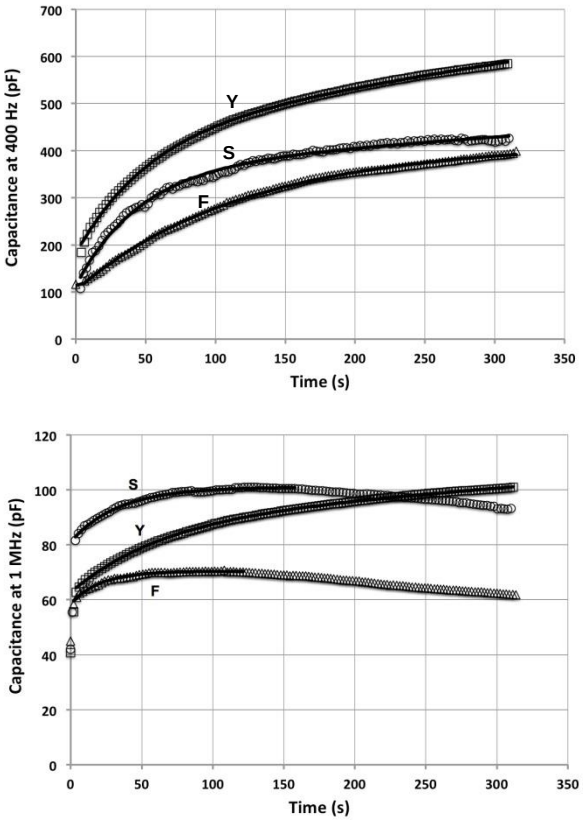

Figure 5. The variation of capacitance with time at (a) 400 Hz; (b) 1 MHz for three subjects: squares: Y (male, 23 years); triangles: F (female, 37 years); circles: S (male, 69 years). Open symbols represent the measured values; solid lines, the fitted values using Eq. 11. Note different data ranges at 400 Hz and 1MHz.

Figure 5(b) shows the comparable results for 1 MHz. The capacitances are now smaller, corresponding to greater penetration depth of the signal [12] and smaller dielectric constant [24]. Note that except for subject Y the capacitances begin to slowly decrease after about two minutes. Such decrease never occurs at 400 Hz and never for subject Y. The cause of these decreases is not clear. The fits using Equation 11 are performed for data up to the beginning of the decrease. The greatest error for the initial data point here is 2.7%. The average difference for the remaining points in the three fits is at most 0.29%. The curves for subjects F and S do not show the long-term viscoelastic response as their capacitances become constant and then decrease after about 100 s.

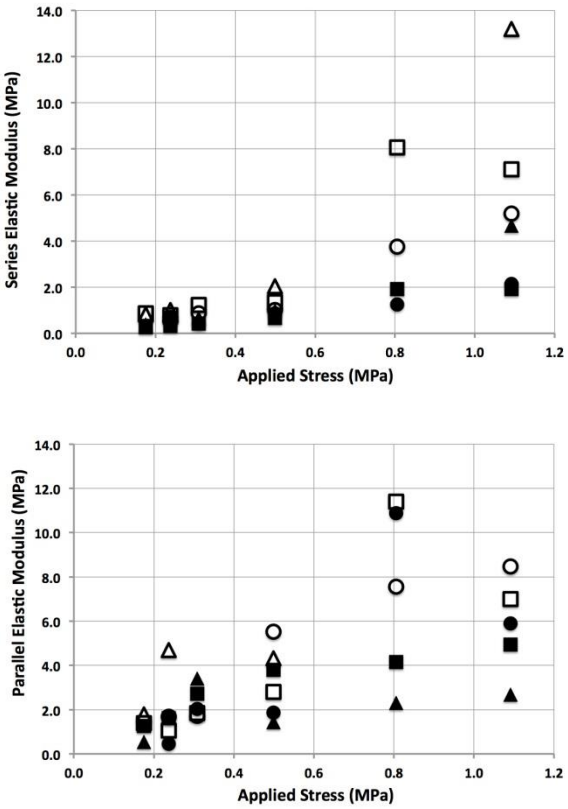

Figure 6. The variation of the elastic moduli with applied stress for three subjects: squares: Y (male, 23 years); triangles: F (female, 37 years); circles: S (male, 69 years). Solid symbols represent values at 400 Hz; open symbols, at 1 MHz; (a) series elastic modulus, E_1; (b) parallel elastic modulus, E_2.

With C_0, C_1, C_2, τ and k determined, the viscoelastic parameters can be found using Equations 5, 6, 8 and 9. Preliminary measurements had shown that the results are independent of position and electrode orientation along the forearm. The elastic moduli obtained from our fits are shown in Figure 6(a) and (b). Figure 6(a) shows the variation of E_1 with the applied stress for the three subjects at the two frequencies. For all three subjects this purely elastic parameter varies non-linearly with the applied stress and is several times larger at 1 MHz than at 400 Hz. A few data points are missing for the F and S results. In those cases the capacitance-time curves are irregular. Figure 6(b) shows similar, nonlinear behavior for E_2, the elastic part of the viscoelastic element. E_2 varies non-linearly with the applied stress for all three subjects. There is a tendency for E_2 to be greater at 1 MHz than at 400 Hz, but it is not as strong as for E_1. However, E_1 is smaller than E_2, except at the highest stress.

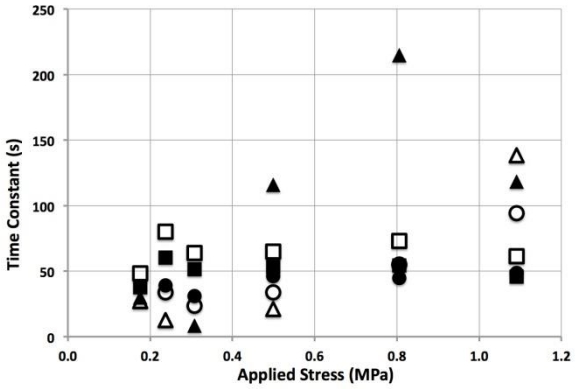

Figure 7. The variation of the viscoelastic time constant, τ, with applied stress for three subjects: squares: Y (male, 23 years); triangles: F (female, 37 years); circles: S (male, 69 years). Solid symbols represent values at 400 Hz; open symbols, at 1 MHz.

As shown in Figure 7, the viscoelastic time constant, τ, does not exhibit any consistent dependence on either the applied stress or the frequency although subject F shows some variation at 400 Hz. These relaxation times, which are on the order of one minute, are apparent because of the lengthy measurement time of about five minutes.

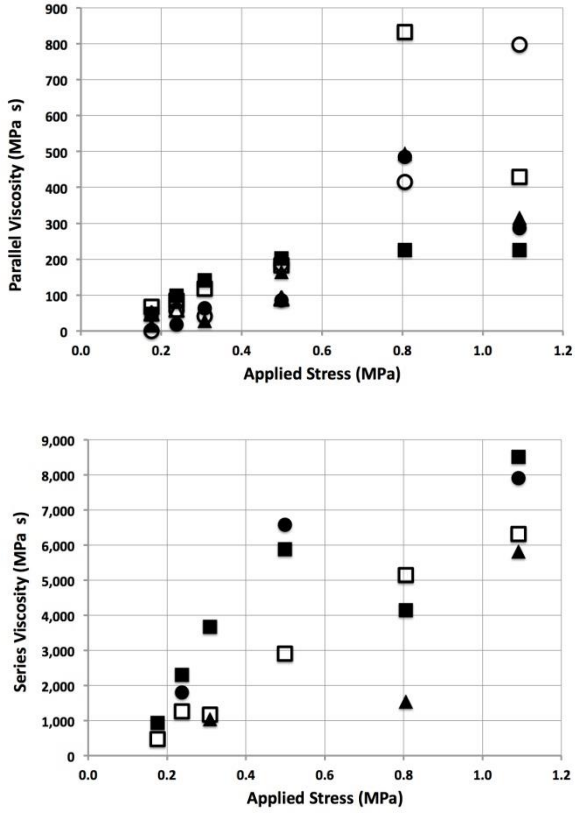

Figure 8. The variation of the viscosity parameters with applied stress for three subjects: squares: Y (male, 23 years); triangles: F (female, 37 years); circles: S (male, 69 years). Solid symbols represent values at 400 Hz; open symbols, at 1 MHz; (a) parallel viscosity parameter, η_2; (b) series viscosity parameter, η_3.

Figure 8(a) illustrates the variation of the parallel viscosity parameter, $\eta_2=\tau E_2$. These viscosities are not those of the interstitial fluid itself, which presumably flows as a result of the compression [3], [19], [27], but are a parameter for the tissue as a whole. As the product of the viscoelastic relaxation time and the elasticity, this parameter also increases with the applied stress, but in a more linear manner, and does not show a consistent difference between the 400 Hz and 1 MHz values. The series viscous element, η_3, also increases with the applied stress as shown in Figure 8(b). Note that for the F and S subjects this element appeared in only a few fits and that the 400 Hz values are consistently larger than the 1 MHz values.

4 DISCUSSION

The aim of our study is to provide a proof of principle for a new method to determine the viscoelastic properties of the skin using measured capacitance changes over time. The capacitances at 1 MHz are significantly lower than those at 400 Hz. According to Equation 1 the capacitance should depend inversely on the effective plate separation, d, and directly on the material relative permittivity, κ. Since the electric current penetrates to a greater depth at 1 MHz compared to 400 Hz [12], and the relative permittivity is smaller at 1 MHz compared to 400 Hz [24], the smaller capacitances at 1 MHz are expected.

In comparing our results with those of other groups several factors must be considered. First, the viscoelastic properties of the skin are nonlinear. The experimental values obtained for the various parameters depend on the applied stress [3]. Second, the results depend on the time frame over which the stress is applied. Long compressions can reveal most completely the viscoelastic behavior. Short compressions on the order of a few seconds or less may reveal only the initial elastic response. To adequately determine the viscoelastic response, the duration of the applied force must be of magnitude comparable to the viscoelastic response time, τ. Alternating stresses for which the compressive phase lasts for half the period will emphasize only the elastic response for frequencies in the Hertz range and underestimate the contribution of the viscoelasticity. Third, some analyses assume a purely elastic response, but note that the response depends on the rate at which the force is applied. This dependence indicates that viscoelasticity is beginning to affect the results, thus reducing their accuracy.

Nevertheless, results obtained in previous indentation experiments are comparable in magnitude to those reported in this paper when the differences in applied stress are taken into account. Tregear and Dirnhuber [19] obtained compression-time curves that are similar to those of Figure 2(a) for times up to 10 minutes. For any particular time the compression was smaller for a greater applied stress, implying an increase of elastic modulus with applied stress. Tregear later confirmed this nonlinearity with a stress-strain plot [28]. He attributed the long-term viscoelastic response to the flow of liquid from the ground substance fluid rather than being a response of the collagen and elastin fibers. Lindahl [20] measured viscoelastic time constants, but not elasticities. He reported two time constants for skin. The first, on the order of 1 s, is faster than could be measured with our system; the second, on the order of 100 s is comparable to the results reported in Figure 7.

Our applied stresses are larger than those reported in other in vivo indentation measurements on skin. Because of the nonlinearity of the viscoelastic response with applied stress, our values must be scaled accordingly for comparison. Moreover, the depth of skin sampled in our method varies with the measuring frequency. Using a frequency below 400 Hz would presumably also yield lower elasticity values. Pailler-Mattel et al. [13], obtained values for skin elasticity lower than ours. Their model assumes two skin layers, dermis and hypodermis, resting on muscle for which the reduced Young's moduli were 35, 2 and 80 kPa, respectively. The authors report forces on the order of 10 mN and contact radii on the order of 1 mm so that the applied stresses are on the order of 30 kPa. Given that their applied stresses are about 100 times smaller than ours along with the nonlinearity of the elasticity, it is not surprising that their elasticity results are also about 100 times smaller.

Jachowicz et al. [21] applied a constant force of 0.06 N for up to 30 seconds on human skin and then used a three-element linear model to analyze the data. Values for E_1, E_2 and η_2 were on the order of 10 kPa, 70 kPa and 100 kPa with considerable individual variability. We estimate that their applied stress was on the order of 1 kPa so that, again, a smaller applied stress yielded smaller values for E_1.

Boyer et al. [22] used a two-element, parallel, linear model to analyze a sinusoidal, 10 Hz indentation. The elastic modulus was on the order of 10 kPa and the viscous modulus was on the order of 1 kPa. We estimate that their applied stress was on the order of 1 kPa. Because the duration of their compressions was much less than a second, the full impact of the viscous element was not measured. In this case the viscous modulus was about 0.1 that of the elastic modulus. In contrast Jachowicz et al. [21] applied the compression for up to 30 seconds and found, for about the same applied stress as Boyer et al. [22] that the viscous element was much greater than the elastic element. Clearly, the relative importance of the measured viscous element depends on the time frame over which the stress is applied.

Iivarnen et al. [23] used an indentor with a 1 mm radius and 4 N force to produce an applied stress on the order of 1 MPa. The resulting elastic modulus value was 0.13 MPa which is smaller than our 400 Hz result in Figure 6(a) for the series elastic element.

In summary, our series elastic parameter, E_1, varied between less than 0.2 to about 10 MPa, as shown in Figure 6(a) for applied stresses between 0.2 and 1.1 MPa. This element was measured over the initial compression on a time frame of a few seconds. Generally, greater stresses yielded greater elastic moduli and much smaller stresses produced much smaller elastic moduli. Our parallel elastic parameter, E_2, was generally somewhat greater than E_1. Jachowicz et al. [21] found a similar relationship although their values were smaller than ours because of their smaller applied stresses.

Our parallel viscosity parameter varied between less than 50 MPa to more than 500 MPa (Figure 8(a)) for applied stresses between 0.2 and 1.1 MPa. Greater stresses yielded greater viscosities. The duration of our compression was 300 seconds. Two groups used an applied stress 0.01 to 0.001 times smaller. Jachowicz et al. [21] found a modulus of about 100 kPa with a duration of 30 seconds whereas Boyer et al. [22] found a modulus of about 1 kPa with a duration much less than a second. Given the differences in applied stress and duration of compression we consider our values to be consistent with previous work. There are no published comparisons for the viscosity of the series viscous element although Tregear and Dirnhuber [19] (1965) and Lindahl [20] obtained a time-dependence similar to ours.

Because our strains are not measured directly, interpretation of our results in terms of skin layers requires careful consideration. The current does not flow along a unique, well-defined depth, but is distributed throughout the dermis and the lower epidermis after passage through the upper epidermis. The viscous response is generally related to the ground substance of the dermis [3], [4], [19], which is a complex, extracellular material containing water, electrolytes and glycoproteins. The viscosity is related to the flow of water from the pressurized region of the dermis into the surrounding tissue. We can then associate the parallel viscoelastic element with the dermis. The upper epidermis, including the stratum corneum, is generally regarded as primarily an elastic element and may be associated with the series elastic element. Finally, the steady, long-term increase in capacitance that requires the series viscous element appeared regularly in the response of the youngest subject, but only occasionally in the other two subjects. It may thus be related to the expulsion of water in the soft, subcutaneous tissue.

It might be argued that this water flow is compromising our electrical measurements. That would be the case if we were using resistance changes for which water transport can be a significant factor. However, we used capacitance changes for our strain determination and capacitance is relatively insensitive to water transport as noted above [8], [9].

This interpretation of our results is consistent with previously published results [13], [21] that indicate skin becomes stiffer with increasing depth. Our E_2 values are greater than our E_1 values and the viscoelastic element is deeper than the purely elastic element. Moreover, both parameters tend to be greater at 1 MHz than at 400 Hz as the higher frequency samples deeper skin layers [2], [12].

5 Conclusion

Our newly developed method provides values for the viscoelastic properties of the skin comparable to those reported in the literature when the dependences on applied stress and duration of compression are taken into account. This agreement with previous results confirms the accuracy of the method. Moreover, our method is well-suited for clinical applications. Indentation is the mechanical stress most commonly experienced by the skin and does not require procedures such as stretching or suction that may cause some discomfort to the patient. Furthermore, complex modeling to obtain the parameters is not needed.

Two aspects of the system provide considerable flexibility in its use. First, we used probes with spring constants that provided relatively large stresses. By changing the spring attachment one can apply a wide range of stresses. Second, because the penetration depth of the electrical signal varies with frequency, one can study the viscoelastic properties of tissue as a function of depth.

Although we have demonstrated the feasibility of this method here for skin, it has a much wider applicability. The ability to determine both the electrical and viscoelastic properties of tissue with the same set of measurements is a definite advantage. For example, an endoscopic version of the probe would permit the analyses of internal tissues, in vivo. The measurement of the viscoelastic properties of cervical [29] and prostate [14] tissues are commonly done on excised samples whereas measurement of the electrical properties can be done in vivo for both cervical [30] and prostate [31] tissues. Our method would permit the measurement of both sets of properties. This combination might provide a more accurate clinical assessment of the tissue's health than either method separately. Measurement of the anisotropic viscoelastic properties of skeletal muscle has been performed on excised tissue [32] whereas the anisotropic electrical properties of skeletal muscle was measured in vivo [33]. The measurement of anisotropic, tissue viscoelastic properties for surgical modeling might be more readily performed using our proposed technique.

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