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Capturing skin properties from dynamic mechanical analyses

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Background/purpose: Existing skin mechanical testing devices focus on measuring skin elasticity and are not tailored to assess the dynamic behavior of skin. Additionally, the mathematical techniques used in existing devices are often not optimal.

Methods: A new dynamic mechanical device that measures the linear dynamics of skin was developed and tested. The mechanical properties of skin were evaluated in experiments in which stiffness and damping parameters were measured (i) at different locations on the arm and hand, (ii) when stratum corneum hydration was varied by controlled changes in environmental humidity, and (iii) following the application of film-forming polymers. Parallel measurements were made with the Cutometer® so that the two devices could be compared.

Results: The findings revealed that reliable and valid measurements of skin mechanical properties can be obtained from the device. The stiffness of the skin was shown to vary significantly as a function of skin site, changes in stratum corneum hydration and following application of the polymer films. Changes in the damping parameter were less consistently associated with varying the condition of the skin.

Conclusion: The high reliability and speed of measurement make this device and analytic procedure an attractive option for testing skin mechanics.

Key words: linear dynamics – stratum corneum – hydration – skin stiffness

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A VARIETY of instruments can be used to measure the mechanical properties of human skin *in vivo* including durometers, skin rheometers, and indentometers, all of which provide information about the mechanical properties of skin (1, 2). These devices employ a range of techniques such as suction, indentation, torsion, extension, ballistometry, and wave propagation to measure skin mechanics *in vivo*. Typically stress–strain relations and measurements of creep and stress relaxation times are measured as a probe indents the skin at a fixed velocity or force or as the skin is lifted, stretched, and released (3, 4). As the pressure or torque of the device increases, the skin first displaces elastically and then creeps once it enters the viscoelastic region. When the system becomes stationary, the device is usually timed to release the pressure or torque, after which the skin relaxes. The application of pressure or torque will normally cause the tissue to have some long-term deformation, which means that the skin does not return to its original state for some time (5).

Devices such as the Cutometer® (Courage and Khazaka, Köln, Germany) and the Dermaflex (Cortex Technology, Hadsund, Denmark) use a suction mechanism in which a pump applies a constant negative pressure at the probe head. The skin in contact with the probe is pulled up into the probe, and sensors mounted in the head of the probe measure the maximum displacement of the skin. This process is inherently nonlinear in that a linear increase in pressure does not result in a proportional increase in the displacement of the skin. Various displacement parameters that represent the elastic, viscoelastic, relaxation, and total displacement properties of the skin are typically calculated from the data sampled. The measurements have also been shown to vary as the number of cycles increased due to progressive creep (4).

Existing skin mechanical testing devices focus specifically on measuring skin elasticity and are

not tailored to assess the dynamic or nonlinear behavior of skin. In addition, the mathematical system identification technique used is often not optimal. It relies on simple step responses that can theoretically contain a lot of important information that can be described using Burger, Maxwell, or Kelvin–Voigt models (6). The simplification of these important parameters into simple displacement values results in a loss of important dynamic information. Models, which often do not contain all the necessary dynamics, are then fitted to experimental curves. A more advanced technique that casts the information into parameters such as stiffness, damping, or energy storage/loss is needed.

A device that can characterize the linear dynamic properties of skin and underlying tissue has considerable potential in cosmetology and dermatology, where it is essential to describe quantitatively the changes in the mechanical properties of skin associated with a treatment or intervention. None of the existing skin mechanical testing devices can fully characterize the dynamic properties of skin, a highly dynamic, nonlinear material. A device that can apply both normal and tangential forces to the skin has been designed and fabricated in the Massachusetts Institute of Technology's (MIT) BioInstrumentation Laboratory; it can characterize the dynamic behavior of skin *in vivo* from data acquired in only 2–5 s (7). The objective of the present set of experiments was to evaluate the reliability and validity of the device in characterizing the mechanical properties of human skin at five locations on the arm under normal conditions, and to measure its performance in detecting changes in skin mechanical properties when the hydration of the skin was changed and following the application of film-forming polymers used in skin care products.

Materials and Methods

Subjects

In Experiment 1, eight subjects were tested. Four were male and four were female, and their ages ranged from 19 to 55 years. In Experiment 2, eight female subjects were tested. Their ages ranged from 19 to 25 years. In Experiment 3, seven female subjects were tested. Their ages ranged from 19 to 23 years. All subjects were healthy and did not have any neurological or dermatological condition that would have affected the skin. Each subject gave informed consent, and all research was approved by MIT's Institutional Review Board.

Instrumentation

Dynamic mechanical device

The device comprises a custom-built Lorentz force actuator that has an inner diameter of 25 mm. It contains a custom-wound coil with a resistance of 14 Ω and was designed as an overhung configuration, that is, the coil windings extended beyond the height of the magnetic field gap. The actuator and probe are housed in a custom fit case fabricated using stereolithography (see Fig. 1) that is mounted on an adjustable aluminum stand. The actuator has a stroke of approximately 30 mm, which is limited by a spring that connects the edge of the probe to the end of the device. The spring acts to push the probe across the skin during the measurement period and ensures that a relatively large area of skin can be tested. The base of the probe has a contact area of 5 by 12 mm. The probe is not anchored to the skin but indents it with a normal force of approximately 1 N and is then moved tangentially across the surface. An ALPS linear potentiometer (model RDC10320RB) is

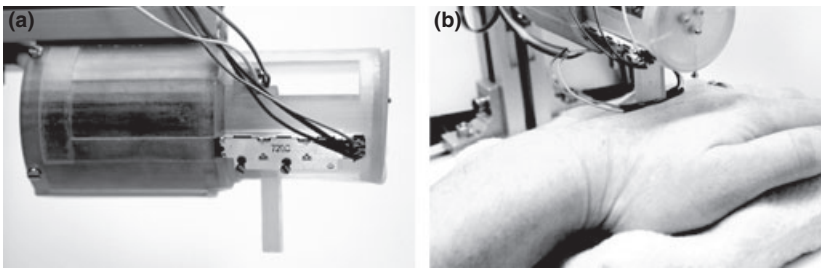

Fig. 1. (a) Device for skin mechanics testing. (b) Device on the dorsal surface of hand during testing.

attached to the coil to measure position. A Honeywell miniature force sensor (model FSS1500NS) is attached to the tip of the probe that makes contact with the skin to ensure that the normal force between the probe and the skin is within the specified range (1.2–1.5 N) at the beginning of the experiment. If the force is not within this range the height of the device is adjusted until the force is correct. Position and force are sampled by 16-bit resolution analog-to-digital converters (ADCs). The coil is driven by a 48 V power supply, and a 5 V power supply provides power to the sensors. With two separate power sources, power supply noise is reduced and the signal quality provided to all parts of the device is increased. Further details regarding the device are available in Chen (8).

Cutometer®

The Cutometer® MPA 580 measures the mechanical properties of skin using suction and is often regarded as the 'gold standard' against which other skin mechanical testing devices are compared (9). The device creates a negative pressure and draws the skin into the aperture of the probe. The probe contains a noncontact optical measuring system that measures the penetration depth of the skin. The aperture diameter of the probe used for these experiments was 2 mm. The resistance of the skin to suction and its ability to return to its original position are given at the end of each measurement. The output parameters include elastic deformation, retraction, viscoelasticity, and ratios involving each of these. In the present experiments the following parameters were calculated: R0: U_f, the elastic deformation of the skin, R5: U_r/U_e, the pure elasticity without viscous deformation, and R6: U_v/U_e, ratio of viscoelastic to elastic extension.

Humidity chamber

A humidity chamber measuring 30 cm × 30 cm × 40 cm was made from acrylic and connected to a humidifier (Electro-Tech Systems Model 572, Glenside, PA, USA). To speed up the humidifying process, a dish of water was placed in the chamber. To decrease humidity, a section of the chamber was left open to the ambient. The chamber was humidified to three levels: 55%, 75%, and 85% relative humidity (RH). The humidity of the chamber was measured using a Vernier relative humidity sensor and recorded using LabVIEW 10.0 (National Instruments Corp., Austin, TX, USA). The humidity of the chamber was monitored during the entire period the arm was enclosed to ensure conditions remained constant. RH was maintained within ± 1.5% of the specified level.

Products

Four gel formulations were tested, three of which contained high molecular weight water-soluble polymers, either polyimide-1 under the name Aquaflex XL-30 or a polyvinylpyrrolidone/acrylate/lauryl methacrylate copolymer under the name of Styleze 2000 (Ashland Specialty Ingredients, Wayne, NJ, USA). These skin-tightening polymers were formulated according to the procedures described by Jachowicz et al. (10). The concentrations of tightening agents were 1% Aquaflex, 3% Aquaflex, and 3% Styleze. The fourth formulation contained no film-forming polymers and was used as a control to ensure that the carrier gel was not contributing to changes in the skin's mechanical properties.

General experimental protocol

The skin on the site being tested was initially placed under the probe. In Experiments 1 (skin site) and 3 (formulations) the normal force exerted by the probe on the skin was maintained between 1.2 and 1.5 N, which resulted in the skin being indented by 1–2 mm. This range of forces was selected based on pilot experiments that showed that the most consistent measurements of stiffness and damping (a coefficient of variation of 5% or less—see Results section) were obtained when the normal force was maintained within this range. In Experiment 2 (hydration) the normal force between the probe and the skin was reduced to between 1.0 and 1.2 N and the probe was coated with a Teflon film in order to facilitate movement of the probe across the hydrated skin. In all the experiments calibrations were performed at the beginning of each set of measurements to measure the contact force of the probe on the skin. If necessary, the position of the device was adjusted so that the normal force was within the specified range. If a subject moved his/her hand or arm during the experiment the force calibration procedure was repeated.

The tangential forces delivered to the skin by the device consisted of a Gaussian stochastic input with a tailored power spectrum having a cutoff frequency of 200 Hz. This cutoff frequency was chosen because it was well above the natural frequency of the skin under study. The stochastic force input was low-pass filtered to boost lower frequencies where it would be expected that the skin mechanics have a higher force to displacement (compliance) gain. The resulting band-limited stochastic signal is a powerful probe for identifying linear dynamic systems. Each trial lasted 5 s, and eight consecutive measurements were taken at each location. The data were sampled at 2 kHz.

For experiments in which the Cutometer® was used, four measurements were taken at the designated sites. The time/strain mode was used in all the experiments. Each measurement lasted 10 s: in the initial 5 s a constant negative pressure of 400 mbar was applied to the skin, followed by a 5-s relaxation period. The data were sampled at 100 Hz.

In all the experiments, the room temperature was maintained at approximately 20°C and the RH was approximately 45%. Subjects were acclimated to the room prior to testing. During the experiments subjects were seated with their arm and hand supported in a supinated position on a contoured cushion on the laboratory bench.

Experiment 1: mechanical properties of different skin sites

The objective of this experiment was to determine the reliability and validity of the device by measuring the mechanical properties of hairy and glabrous skin and comparing the device's performance to that of the Cutometer®. Five locations on the forearm and hand were selected for study: the posterior surface of the forearm near the wrist, the anterior surface of the forearm near the wrist, the anterior surface of the forearm near the elbow, the dorsal surface of the hand, and the thenar eminence, as shown in Fig. 2.

Experiment 2: effects of epidermal hydration on skin mechanics

The objective of this set of experiments was to measure the mechanical properties of skin as a function of its hydration. Variations in skin hydration were achieved by varying the RH of a chamber in which the arm was immersed. It has been shown that at constant temperature, an increase in the RH of the environment causes an increase in the moisture content of the stratum corneum as demonstrated by a decrease in the skin's electrical impedance (11) or increase in capacitance (12). In the present experiment three settings of RH were used (55%, 70%, and 85%) in addition to the baseline condition measured at an ambient RH of 45%. The baseline measurements with the device and Cutometer® were taken after each subject was equilibrated for 30 min to the ambient conditions. This duration was comparable to those used in other studies of skin hydration (e.g., 9, 11, 12). Immediately following the baseline measurements, the arm was placed in the humidity chamber for 30 min, which was initially set to 55% RH. The same procedure was followed for the measurements taken after immersion of the arm in the chamber with its RH set at 70% and 85%. All measurements in this experiment were taken on the anterior forearm position 2 (see Fig. 2) immediately after the arm was withdrawn from the chamber.

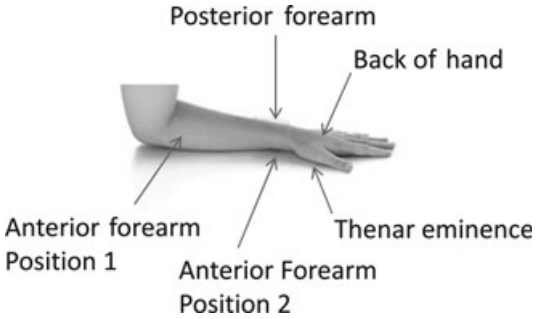

Fig. 2. Diagram showing the five skin sites tested.

Experiment 3: effects of topically applied formulations on mechanical properties of the skin

In this set of experiments the same experimental procedure was used to evaluate the effects of a range of formulations on the stiffness and damping of the skin on the forearm. These high molecular weight polymers are commonly used in the cosmetic industry to induce skin tightness, which is accompanied by skin smoothing and elimination of winkles and lines (10). Measurements were performed on the anterior surface of the forearm under normal conditions (untreated) and after application of one of three formulations. The formulations were applied to a 3.8-cm diameter circle on the skin and distributed to form a continuous film, as described by Jachowicz et al. (13). The films were left on the

skin to dry for 2 min, and then the mechanical properties of the treated skin were tested.

Data analyses

Data acquisition and display were performed in LabVIEW 10.0 and linear system identification was conducted in MATLAB (MathWorks, Natick, MA, USA). Nonparametric compliance impulse response functions, which are a complete description of the linear dynamic relation between the input (force) and output (displacement), were calculated for each trial using a least mean squares method involving Toeplitz matrix inversion. The latter involves deconvolving the input autocorrelation function from the input–output cross-correlation function. The overall system dynamics included both the skin and the actuator dynamics. Before the measurements were made on the skin, calibrations were performed on the device so that its linear properties could be assessed. After the skin measurements were completed, the dynamics of the device were removed, leaving only the skin dynamics.

The nonparametric impulse response functions were well approximated by second-order under-damped low-pass parametric impulse response functions. The general form of the fitted second-order low-pass under-damped impulse response can be obtained from the inverse Laplace transform of the system's compliance transfer function:

$$H(s) = \frac{1}{Is^2 + Bs + K}, or \quad equivalently$$
$$H(s) = \frac{Gain \cdot \omega_n^2}{s^2 + 2\zeta\omega_n s + \omega_n}, \tag{1}$$

where K is the stiffness, B is the viscous stiffness, and I is the inertia. ω_n is the natural frequency, $Gain$ is the static compliance, and ζ is the damping parameter. Note that

$$Gain = \frac{1}{K}, \omega_n = \sqrt{\frac{K}{I}}, \text{ and } \zeta = \frac{B}{2\sqrt{I \cdot K}}$$

or equivalently

$$I = \frac{1}{Gain \cdot {\omega_n}^2}, B = \frac{2\zeta}{Gain \cdot \omega_n} \text{ and } K = \frac{1}{Gain}$$

The inverse Laplace transform of the compliance transfer function, $H(s)$, is the compliance impulse response function, $h(\tau)$. For the case where the system is underdamped ($\zeta < 1$),

$$h(\tau) = Gain \cdot \omega_n \cdot e^{-\zeta\omega_n t} \frac{\sin\sqrt{1-\zeta^2} \cdot \omega_n t}{\sqrt{1-\zeta^2}}. \tag{2}$$

This function was fitted to the calculated nonparametric impulse response functions using a least-squares nonlinear parameter estimation technique (see Fig. 3). The resulting estimates of $Gain$, ω_n, and ζ were converted using the equivalences above to corresponding I, B, and K values. The stiffness estimates refer to the deformation of the skin as a function of the force applied. The damping parameter is dependent on the inertia (I), elastic stiffness (K), and viscous stiffness (B) of the skin, and is dominated by the viscous term. The damping parameter is proportional to viscosity and inverse square root proportional to the inertia and elastic stiffness. In general, the linear second-order under-damped, low-pass model provided a good fit to the experimental data with the variance accounted for by the model always exceeding 80% and for most of the data it exceeded 90%. Statistical analyses were conducted to determine whether differences in the stiffness and damping parameters measured in the various experiments were significant. These analyses were performed using SPSS (IBM, Armonk, NY, USA).

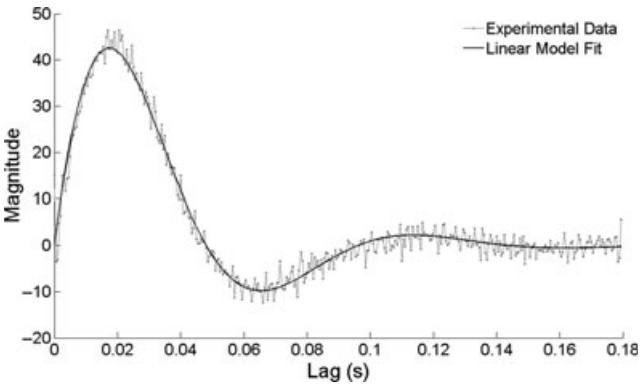

Fig. 3. Compliance impulse response function (of the device and skin) calculated from measurements made on the anterior surface of the forearm. The linear second-order under-damped, low-pass model fitted to the experimental data is shown in black.

Results

Experiment 1: mechanical properties of different skin sites

The mean values measured for the stiffness and damping parameters at each of the five sites

tested are illustrated in Figs 4 and 5. A repeated measures analysis of variance (ANOVA) of these data indicated that there was a significant difference in the stiffness and damping parameters measured at the five sites [stiffness: $F(4, 28) = 23.09$, $P < 0.001$; damping: $F(4, 28) = 12.27$, $P < 0.001$]. The highest stiffness and damping parameter values were measured on the glabrous skin on the thenar eminence of the hand as evident in Figs 4 and 5.

The reliability of the device was evaluated in terms of the coefficient of variation (100·SD/mean), which is a commonly used index of the consistency of a device's performance (14, 15). The mean coefficients of variation (CV) for stiffness and damping were calculated at the five sites tested and the mean CV for the stiffness and damping estimates were 3.2% and 3.7%, respectively. In general, the CVs did not vary much as a function of the site tested and ranged from 2.9% to 3.6% for stiffness and 3.2–4.0% for damping. The sites were also tested using the Cutometer® MPA 580 and the mean CVs for the three parameters frequently calculated with this device (R0, R5, and R6) were 5.7%, 6.2%, and 16.6%, respectively.

The relation between the parameters related to the skin's elasticity (R0 and R5) calculated from measurements made with the Cutometer® and the stiffness measured by the Dynamic Mechanical Device (DMD) were evaluated. The correlation coefficients (Pearson product moment) between the R0 and R5 parameters and the stiffness measured by the DMD were -0.53 and -0.51, respectively. These values indicate there was a modest and significant relation ($P < 0.01$) between the variables measured with the two instruments. The correlation between the R5 parameter and stiffness was significant at $r = -0.39$ ($P < 0.05$).

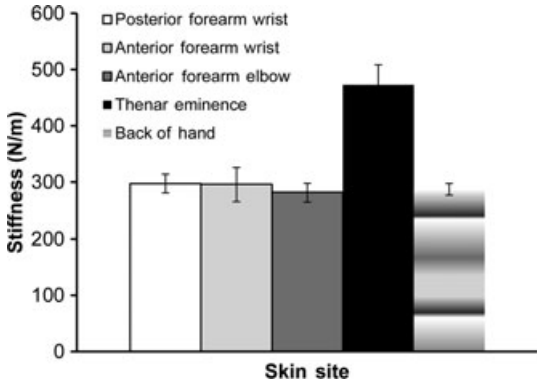

Fig. 4. Group mean skin stiffness measurements (± SEM) at five locations.

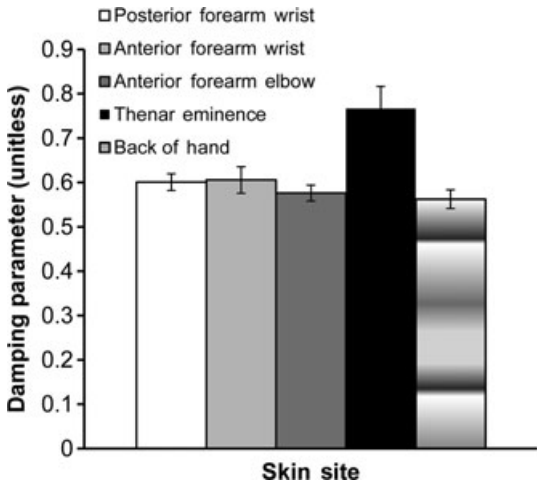

Fig. 5. Group mean skin damping parameter measurements (± SEM) at five locations.

Experiment 2: effects of epidermal hydration

The same general procedure described above was used in the experiments conducted to determine how the skin's stiffness and damping changed as a function of varying the degree of epidermal hydration. The skin's hydration was varied by immersing the arm in a chamber in which the RH was controlled. Barel and Clarys (12) have shown that there is a linear relation between external RH (over the range of 35–85% RH) and the hydration of the skin as measured by its capacitance. As illustrated in Fig. 6, the stiffness increased linearly as the RH increased from 45% to 85%. A repeated measures ANOVA of these data indicated that there was a significant increase in stiffness as the hydration of the skin increased [$F(3, 21) = 9.99$, $P < 0.01$]. In contrast there was no change in the damping parameter measured as the RH increased ($P = 0.9$) (see Fig. 7). Results from measurements made with the Cutometer® are presented in Fig. 8(a–c). The R0 parameter (elastic deformation) increased as the RH increased, consistent with the measurements of stiffness. As evident in Fig. 8(b) and 8(c), there was no consistent trend in the R5 and R6 parameter measurements, although the R5 parameter did decrease linearly until the RH reached 70% and thereafter increased.

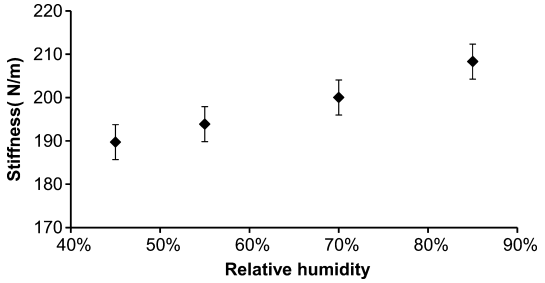

Fig. 6. Group mean stiffness (± SEM) as a function of relative humidity.

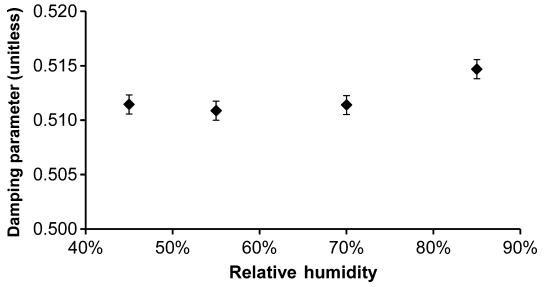

Fig. 7. Group mean damping parameter estimates (± SEM) as a function of relative humidity.

Experiment 3: effects of formulations on mechanical properties of the skin

Of the four formulations applied to the skin, three contained skin-tightening polymers (1% Aquaflex, 3% Aquaflex, and 3% Styleze) and the final formulation did not contain any film forming polymers and was used as a control gel to ensure that the carrier of the stiffening agents did not contribute to the changes in the mechanical properties of the skin. Fig. 9 shows the mean stiffness values measured following application of the various formulations and under normal (untreated) conditions. The values for stiffness of the untreated skin ranged from 209 to 230 N/m, consistent with the results obtained in testing different skin sites (see Fig. 4). Application of the formulated gels to the skin resulted in an increase in its stiffness, with increases ranging from 3.1% to 13.3% (mean 8.6%) for the 1% Aquaflex to 6.2% to 27.4% (mean 13.9%) for the 3% Aquaflex formulation. The 3% Aquaflex and 3% Styleze gels produced average stiffness values of 246 and 244 N/m, respectively, suggesting that their effects on skin stiffness are very similar. The gel without film-forming polymers caused a much smaller change in stiffness that ranged from 0.3% to 1.6%. A repeated-measures ANOVA of these data revealed that there was a significant difference in the stiffness of the skin as a function of the formulation applied [$F(4, 24) = 15.97$, $P < 0.001$].

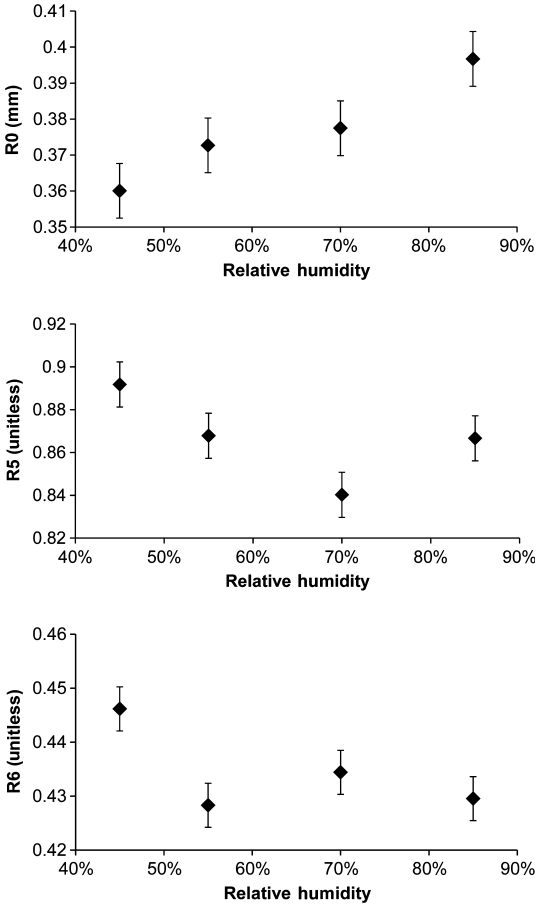

Fig. 8. (a) group mean R0; (b) group mean R5; and (c) group mean R6 parameter estimates from the Cutometer® as a function of relative humidity. The SEM are shown.

Application of the formulations generally resulted in an increase in the damping parameter, but the change was more variable than that found for stiffness. The group mean results illustrated in Fig. 10 show an overall increase in the damping parameter following application of the gel containing Styleze and for both gels containing Aquaflex. However, there was no statistically significant difference in the damping parameter estimates as a function of the formulation applied to the skin ($P = 0.35$). The within-subject damping data were nevertheless quite consistent as reflected in the mean coefficient of variation of 3.3% across all conditions.

Discussion

This series of experiments revealed that the DMD provides reliable estimates of the mechan-

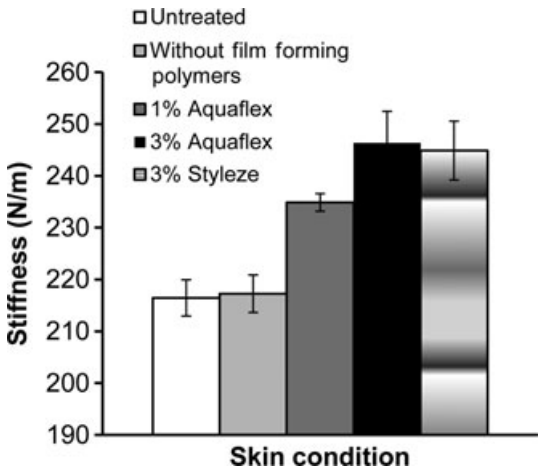

Fig. 9. Group mean stiffness estimates (±SEM) of untreated skin and following the application of various formulations.

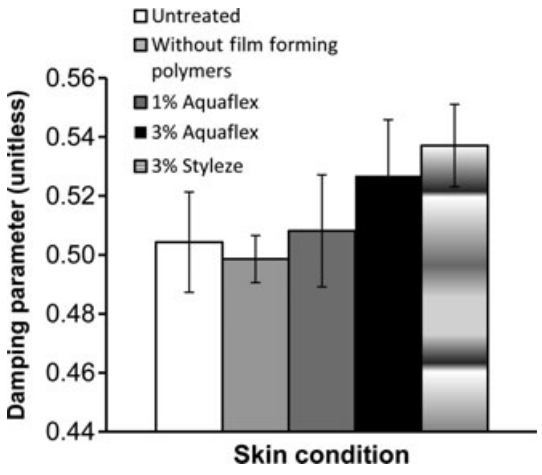

Fig. 10. Group mean damping parameter estimates (±SEM) of untreated skin and following the application of various formulations.

ical properties of skin *in vivo*. Its reliability was measured in terms of the CV and found to average 3.2% for stiffness and 3.7% for the damping parameter. These values compare very favorably with estimates of the reliability of the Cutometer®, which ranged from 5.7% to 16.6% for the parameters measured in the present experiment. Barel et al. (14) reported that the CV for the R0 parameter (total skin distension) measured with the Cutometer® varied between 4% and 6%, which is very similar to the value (5.7%) found in the present experiment. The reproducibility of skin mechanical parameters estimated using other suction devices such as the Echorheometer ranges between 5% and 12.4% (16), and is higher than 10% for the Reviscometer® (15). The DMD was also evaluated in terms of the relation between the parameters it measures and those of the Cutometer®. Modest and significant correlations were found between the parameters measured by the two devices. It was not anticipated that the correlation would be very high due to the differences in the nature of the mechanical inputs delivered to the skin and their principles of operation (suction vs. extension). It appears that the measurements of stiffness and damping using the DMD complement viscoelastic parameters obtained from suction devices such as the Cutometer®.

The mechanical properties of skin primarily depend on the dermal and hypodermal collagen and elastic fiber network that is embedded in a viscous ground matrix, to which the epidermal layer also contributes. In the present experiments the stiffness and damping estimates obtained were very similar for the four areas of hairy skin tested and both mechanical parameters were considerably higher for the glabrous skin on the thenar eminence. In contrast to hairy skin, glabrous skin has a thick epidermis, with a stratum corneum that ranges from 100 to 200 μm, as compared to 10–40 μm in hairy skin. In addition, the skin on the palm of the hand is anchored to the underlying fascial planes by fibrous tracts that prevent the skin from gliding over the underlying tissue (17). These structural differences between the two types of skin contribute to the greater stiffness of glabrous skin.

The stiffness of hairy skin averaged around 300 N/m in Experiment 1, which is comparable to the stiffness measured using a suction device on the forearm (18). Estimates of the mechanical properties of skin such as its Young's Modulus vary considerably (0.02–57 MPa) depending on the model proposed and the stresses applied to the skin (19). External variables such as the force and displacement applied to the skin, and the surface area perturbed by the device, can also have a profound effect on the mechanical parameters measured.

The hydration of the skin was varied by immersing the forearm in a chamber in which the RH was varied and then measuring the mechanical properties of the skin on the forearm. An increase in the RH of an environment at a constant temperature causes an increase in the moisture content of the stratum corneum as demonstrated by an increase in electrical impedance (10). The stiffness of the skin was

found to increase linearly as the hydration level of the skin increased, consistent with other experiments that have shown that hydration of the skin significantly increases its stiffness (3, 9) and the coefficient of static friction (20–22). As the water concentration and distribution in the stratum corneum increases with higher levels of hydration two separate phenomena occur that contribute to the increase in measured stiffness. One is related to the change in surface tension of the liquid between the skin and the mechanical testing device, and the other has been hypothesized to involve solubilization of the protein chains in the cells in the stratum corneum (23). The stiffness measured in this experiment was lower than that measured in Experiment 1. This is probably due to the smaller normal force used, which was required for the probe to move easily across the hydrated skin surface and the absence of any males in the subject pool for the hydration experiment. The measured stiffness of the male subjects was noted to be on average 25% higher than that of the females in Experiment 1.

The mechanical properties of skin can also be changed by applying thin polymer films that are designed to induce skin tightness or firming (10). Consistent with the results from other studies that have used various devices to measure the changes in skin associated with application of these polymer films, in the present experiment application of the formulations resulted in a significant increase in skin stiffness. Moreover, the stiffness of the skin was determined to increase progressively with higher concentrations of tightening agents. The damping parameter also changed following the application of the films, although the change was not statistically significant. As evident in the standard errors shown in Fig. 10, there was considerable variability across subjects in terms of the change in the skin's damping parameter.

Two parameters were selected to characterize the mechanical properties of skin in these experiments, namely stiffness (K) and damping (ζ). The damping parameter is not independent of stiffness, although it is dominated by the viscous term (B). The changes in viscous stiffness (B) generally mirrored the variations in the damping parameter reported for the three experiments. Experiment 1 involving different skin sites was the only experiment in which there was a significant difference in the viscous term, as was also found for the damping parameter. The natural frequency followed a similar pattern in that it ranged from 70 to 80 rad/s in most experiments and was substantially higher (mean: 103 rad/s) on the palmar skin.

The results from this series of experiments indicate that the dynamic mechanical device captures the dynamic mechanical behavior of the skin under normal conditions, when the hydration of the skin is changed with variation in environmental humidity, or following the application of formulations containing film-forming polymers. The device was shown to be reliable and capable of producing measurements in a very short period of time. It can be used for both *in vitro* and *in vivo* studies of skin, and has not been shown to affect the mechanical properties of skin so it can be used for repeated measurements. Although the device was fixed to an aluminum frame for these experiments it can readily be converted into a relatively low-cost portable system for measuring skin mechanical properties. Future research with this device will focus on augmenting the analytic techniques so that the nonlinear behavior of skin can be characterized and on developing a portable handheld version of the device.

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