INTRODUCTION

The skin is a complex structure with highly non-linear mechanical characteristics [1], which are of interest to researchers in dermatology, haptics, and neurobiology. However, measurement of the skin’s many dimensions remains a complex undertaking. Most prior work has utilized tensile loading techniques [2], [3], under which the skin behaves very different as compared to compression loading, especially for large strains [1], [4], [5]. Further the gap in our understanding of the skin’s compressive behavior extends in particular to its nonlinear, viscoelastic properties. Herein we examine the nonlinear viscoelasticity of the skin under compressive loading, which we control at different displacements and therefore levels of strain.

METHODS

The methods utilize controlled displacement, linearly ramped into the skin surface, at commanded velocity of 100 mm/s to collect time-force-displacement data from which we generate material parameters in the commonly used quasilinear viscoelasticity model [6]. Also calculated was skin thickness, using a new method based on contact force, which contrasts with traditional methods that utilize calipers and glass slides [7]. The measured specimen was obtained using a 6-mm diameter biopsy punch, from skin sites on hind limb of a 9.6-week old mouse, 19.63 grams of body weight. The skin was tested whereby the compressive strain, for one skin specimen, was varied at 10 levels of equal increments (Figure 1A). Data from the 6th run of 10 repetitions at each level of strain was analyzed in order to avoid variance introduced by loading history.

The quasilinear viscoelasticity model uses the Boltzman convolution to calculate stress from strain data,

\[ \sigma(t) = \int_{-\infty}^{t} G(t \to t') \frac{\partial \sigma(\lambda)}{\partial \lambda} dt' \quad (1) \]

where \( t \) and \( \lambda \) denote time and stretch at any given moment. \( \sigma(\lambda) \) denotes the instantaneous elastic function of material. Since a fast loading velocity was adopted, the loading phase in analyzing the strain-dependency was simplified into an ideal step load

\[ \lambda(t) = \begin{cases} 1, & t < 0 \\ \lambda_{\text{min}}, & t \geq 0 \end{cases} \quad (2) \]

where \( \lambda_{\text{min}} \) denotes the stretch under full load. By inserting Eqn. (2) into Eqn. (1) and integrate the Dirac delta function, we attain

\[ \sigma(t) = G(t) \sigma(\lambda_{\text{min}}) \quad (3) \]

Note that \( \sigma_{i}(\lambda_{\text{min}}) \) is equivalent to the stress at time zero \( \sigma_{i} \). We also moved \( G(t) \) to the left hand side,

\[ G(t) = \frac{\sigma(t)}{\sigma_{i}} \quad (4) \]

to attain the reduced relaxation function, which has the form of a two-term Prony series as in Eqn. (5),

\[ G(t) = \sum_{i=1}^{2} G_{i} e^{-\frac{t}{\tau_{i}}} + G_{\infty} \quad (5) \]

with unity constraint in Eqn. (6),

\[ \sum_{i=1}^{2} G_{i} + G_{\infty} = 1 \quad (6) \]

Among strain measures such as true strain, Biot strain and Green strain, Green strain was used, as defined in finite deformation,

\[ \varepsilon = \frac{(x^2 - 1)}{2} \quad (7) \]

Force-time data (Figure 1A) from the experiment on strain-dependency were converted to stress-time data by dividing force over sample area, smoothed via cubic spline to attain more accurate peak stress values, and then normalized by the peak force of each trace (Eqn. (4)) before fitting to the Prony series of Eqn. (6). The fitting returned an average \( R^2 \) value of 0.98, with 10 resultant reduced relaxation functions \( G(t) \) as shown in Figure 1B.

VISCOELASTICITY OF MOUSE SKIN UNDER COMPRESSION IS DEPENDENT ON LEVEL OF STRAIN

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RESULTS

As strain increases, before reaching 0.28, we observe greater relaxation at the same time constants, and after 0.28 less relaxation at longer time constants. In specific, two discernible strain regions were observed: region I from a strain of 0.17 to 0.28; region II from a strain of 0.28 to 0.37 (Figure 2A). For example, when level of strain increases from 0.17 to 0.28, the time scale of the relaxation almost remains unchanged but G(t) at 0.28 relaxes more than that of 0.17. Again, when level of strain increases from 0.28 to 0.37 the time scale is postponed while G(t) relaxes less in 0.37 to 0.28. With fitted parameters we note that in region I, the two time constants (τ₁ and τ₂) remained steady (Figure 2C) as G₁ increased with level of strain and G₂ decreased with level of strain (Figure 2B). In region II, both time constants increased with level of strain (Figure 2C) while G₁ decreased, G₂ remained constant and Gₓ increased with level of strain (Figure 2B). Each increase and decrease passed Pearson test of correlation, as listed in Table 1.

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REFERENCES


Table 1 Correlation coefficients between level of strain and viscoelastic parameters τ and G₁ in region I and II.

<table>
<thead>
<tr>
<th>Region</th>
<th>τ₁</th>
<th>τ₂</th>
<th>G₁</th>
<th>G₂</th>
<th>Gₓ</th>
</tr>
</thead>
<tbody>
<tr>
<td>Region I</td>
<td>0.40</td>
<td>-0.32</td>
<td>0.85</td>
<td>0.99</td>
<td>-0.90</td>
</tr>
<tr>
<td>Region II</td>
<td>0.83</td>
<td>0.96</td>
<td>-0.90</td>
<td>0.52***</td>
<td>0.92</td>
</tr>
</tbody>
</table>

*** denotes p-value > 0.5; ** denotes 0.1 < p-value < 0.5; * denotes 0.05 < p-value < 0.1; others p-value < 0.05. Grey-scale shading is used to denote positive/negative/neutral correlation.

DISCUSSION

Work herein shows dependency of skin viscoelasticity on levels of strain. One alternative biological reasoning is the viscoelasticity of skin owes itself to two characteristics of its components, the viscosity of interstitial fluid [8], and complex matrix structure of collagen and elastin solid [1]. These two characteristics can be mapped to the two-region behavior we observed upon varying the level of strain: 1) when finite deformation begins (region I in Figure 2B and C) the interstitial fluid viscosity dominates since it is mostly free fluid movement, 2) at a later point in the indentation (strain = 0.28 in this case) the matrix structure of elastin, and to some extent collagen, dominate a more typical solid elasticity (region II in Figure 2B and C).

We note that phenomenon reported herein was from analysis of one representative sample, while other skin specimens behaved in similar fashion. The results also suggest the need for modifications to the current quasi-linear viscoelastic model where values of G₁ and τ₁ are linear with level of strain, compared to being constant.

Figure 1 A: Force traces under ten levels of strain. B: Fitted reduced relaxation functions under the ten levels of strain, three examples of which are denoted on the right (17%, 28% and 37% strain). Note that five pre-indentation traces are not shown.

Figure 2 A: Three typical reduced relaxation functions. B: Values of G₁ when level of strain varies. C: Values of time constants when level of strain varies.