Laser Speckle Strain Measurements in Soft Tissue

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Abstract

The importance of understanding the mechanical behavior of tissues is starting to be recognized in medical diagnostics. It is becoming clear that pathological tissues behave mechanically differently than do healthy tissues. However, traditional optical evaluation methods, particularly speckle-based methods, have not proven to be effective in assessing the mechanics of biological soft tissues. This is due, in part, to the rapid temporal decorrelation of speckle patterns and the depolarization of the scattered light, making fringe generation and interpretation extremely difficult. A laser-speckle strain sensor will be described that has been shown to be effective in high-resolution strain and strain rate measurements in fully hydrated biological hard and soft tissues. The sensor does not rely upon fringe generation as do most speckle-based methods, but instead relies upon high-resolution speckle tracking based on a maximum likelihood shift estimator coupled with a differential illumination arrangement that makes the sensor sensitive only to strains in the plane of the illumination and observation. Herein, we will focus on a variation of the strain sensor that can be used to quantify certain viscoelastic parameters of soft biological tissues. The ability of the sensor to differentiate between healthy and chemically altered tissues will also be discussed.

Introduction

Laser speckle strain or displacement measurements in soft tissue are often confounded by the very rapid decorrelation of the speckle patterns due to tissue and water movement, as well as the depolarization of the speckle fields. This decorrelation is largely a result of the tissue behaving as a volume scatterer as opposed to a surface scatterer, such as lightly abraded metals [1]. Furthermore, different wavelengths of light will penetrate to different depths within the tissue, thus different wavelengths of light have different probing depths through which the measurement is made. Because of these issues, classical speckle interferometric methods are often not suitable for mechanical measurements in fully hydrated tissues. Herein, we present a variation on a laser speckle strain gauge that does not rely upon the formation of correlation fringes for displacement measurements. The gauge is relatively insensitive to modest speckle decorrelation, has no dependence on the polarization state of the scattered light, and is highly sensitive (sub-microstrain resolution). The variation that we describe is a system that has been shown to be effective in quantifying aspects of the viscoelastic behavior of soft tissues [2,3]. This system has potential applications in medical diagnostics, as well as basic biomechanical studies.

The strain gauge we describe is based on the laser speckle strain gauge of Yamaguchi [4]. Duncan \textit{et al.}, [5] reported on variations of Yamaguchi's strain gauge, with the primary difference being in the manner in which the speckle data were processed, leading to improved strain sensitivity. In these papers, the concept of a 'stacked speckle history' was introduced. Stacked speckle histories are sequences of 1-dimensional views (images if subjective speckle is being employed) of the backscattered speckle pattern from an object stacked into a 2-dimensional array such that pixel number of the camera (space) is along the abscissa and record number (or time) is along the ordinate. If the object is straining and, in the case of subjective speckle, if there is a slight mis-focusing of the imaging system, then the individual speckles will translate in both space and time so that the stacked speckle histories take on the appearance of a tilted corrugated structure, with the tilt relating to the displacement rate of the speckle pattern and the corrugation representing the phase of the individual speckles.

Yamaguchi [4] and Duncan \textit{et al.}, [5] showed that for an object straining in the \textit{x-y} plane, illuminated through an angle $\theta_s$ in the \textit{x-z} plane, and observed normal to the object surface (i.e., in the \textit{z}-direction), the speckle displacement $\delta \hat{x}(0, \theta_s)$, is given by

$$
\delta \hat{x}(0, \theta_s) = a_x \left[ \frac{L_x \cos \theta_s - \cos \theta_o}{L_x \cos \theta_s + \cos \theta_o} \right] - a_x \left[ \frac{L_x \cos \theta_s \sin \theta_s - \sin \theta_o}{L_x \cos \theta_s + \sin \theta_o} \right] - L_o \left[ \frac{\sin \theta_o + \tan \theta_s}{\cos \theta_s} \right] - \Omega \left[ \frac{\cos \theta_s \sin \theta_o + 1}{\cos \theta_s} \right].
$$

\text{(1)}
where \( a_i \) is the rigid body motion in the direction indicated by the indices, \( L_o \) is the observation distance, \( L_s \) is the source distance, and \( \Omega \) is the rotation about the \( y \)-axis. The desired term, that is, the in-plane component of the strain tensor, \( \varepsilon_{xx} \), can be isolated by using collimated source beams (i.e., \( L_s \to \infty \)) and sequentially illuminating the object through two equal, but opposite beams, \( \pm \theta \). Subtracting the speckle motions as seen for the two illumination angles results in a simple expression for the differential speckle motion, \( \delta A \):

\[
\delta A = \delta x(0, +\theta) - \delta x(0, -\theta) = -2L_o \varepsilon_{xx} \sin \theta.
\]  

For purposes of computational efficiency that will become apparent below, it is advantageous to take the time derivative of Eq. (2). Doing so and re-arranging yields:

\[
\dot{\varepsilon}_{xx} = \frac{\delta \dot{x}(0, +\theta) - \delta \dot{x}(0, -\theta)}{2L_o \sin \theta},
\]  

where the time derivative is indicated explicitly by the dot over the character. Thus, the strain gauge is sensitive only to the in-plane component of the 3-D strain tensor.

The key to the strain gauge is the efficient and accurate estimation of the speckle shift between successive records as a result of an imposed load. A variety of both nonparametric and parametric estimators have been investigated in the past [1]. These include cross-correlation, the transform method of processing laser speckle data, maximum-likelihood speckle shift estimators, and maximum entropy speckle shift estimators. For large speckle shifts (>0.5 pixels/record), any of the estimators perform satisfactorily [1], however, for small shifts (<0.5 pixels/record), the maximum likelihood estimator has been shown to be superior to the other approaches. Duncan and Kirkpatrick [6] presented a numerical performance analysis of the maximum likelihood estimator and demonstrated that this shift estimator performs robustly in the presence of noise. Furthermore, it was shown that by employing an iterative processing routine, the estimator performs very well (i.e., the estimated shift closely approximated the actual, imposed shift) for shifts ranging from 0.1 pixels/record to 0.8 pixels/record. Shifts below 0.1 pixels/record were not investigated.

Variations on the laser speckle strain gauge described herein and the processing schemes referenced above have been used in the past to evaluate the micro-mechanical behavior of steel billets [5], fine wires [7], orthodontic wires [8] bovine cortical bone [9], and human vascular tissue [10], to provide a few examples.

Methods and Materials

Speckle Tracking Algorithm

The details of the numerical algorithm used to estimate the speckle pattern shift as a function of the imposed load have been given elsewhere [1,6], so only the key features of the algorithm will be presented herein. The crux of the method is to determine the shift, \( \zeta \), of the speckle pattern as a function of record number (time) and applied load. This shift was calculated using a maximum likelihood approach developed in earlier works. We adopted a “frozen speckle” model, much like Taylor’s frozen turbulence hypothesis and assume that, over a time on the order of a couple sequential exposures of the camera, the structure of the speckle pattern is fixed. The only change with time is its lateral motion. Thus, the speckle motion can be modeled as

\[
g_{j+1}(x_i) = g_j(x_i - \zeta),
\]  

where the subscript \( i \) denotes the pixel (spatial dimension) and the subscript \( j \) represents the record (temporal dimension). We assume that the shift, \( \zeta \), is small compared to a pixel so that Eq. 4 can be approximated as

\[
g_{j+1}(x_i) \approx g_j(x_i) - \zeta \cdot g'(x_i).
\]  

This is simply the first two terms of the Taylor series expansion for \( g \). To introduce a degree of symmetry into the problem, we inspect the two speckle records on either side of the record of interest,

\[
\{g_{j+1}(x_i), g_{j-1}(x_i)\}.
\]  

We then determine the \( \zeta \) that minimizes the error,

\[
e^2_j = \sum_{i=1}^{N}\left[ g_{j+1}(x_i + \zeta) - g_{j-1}(x_i - \zeta) \right]^2,
\]  

where the summation is over all pixels in the two records. We are thus seeking the \( \zeta \) that brings these two records into registration. If we make use of the approximation in Eq. 5, then differentiating with respect to \( \zeta \) and rearranging yields the formula
\[
\hat{\zeta}_j = \frac{-\sum_{i=1}^{N} \left[ g_{j+1}(x_i) - g_{j-1}(x_i) \right] \left[ g'_{j+1}(x_i) + g'_{j-1}(x_i) \right]}{\sum_{i=1}^{N} \left[ g'_{j+1}(x_i) + g'_{j-1}(x_i) \right]^2}.
\]

(8)

The term in the first square bracket in the numerator is proportional to the first central difference approximation to the derivative;

\[
\frac{\hat{g}_{j}(x)}{\hat{t}_{j}} \approx \frac{g_{j+1}(x) - g_{j-1}(x)}{2}.
\]

(9)

The spatial derivatives may be approximated similarly:

\[
\frac{\hat{g}_{j+1}(x)}{\hat{x}_{j}} \approx \frac{g_{j+1}(x) - g_{j-1}(x)}{2},
\]

\[
\frac{\hat{g}_{j-1}(x)}{\hat{x}_{j}} \approx \frac{g_{j-1}(x) - g_{j+1}(x)}{2}.
\]

(10)

Note that the speckle shift parameter, \(\zeta\), is the time rate at which the speckle pattern shifts. The units are pixels/record. This estimator was derived in terms of the shift that minimizes the mean-square error (MMSE) between adjacent records. It is also the maximum likelihood (ML) estimate of the shift [1,6]. Note that the same arguments as above can readily be expanded into evaluating full 2-D image stacks.

**Evaluation of tan \(\delta\) in porcine skin**

The remainder of this paper focuses on a variation of the strain gauge described above to evaluate certain viscoelastic properties of porcine skin, in particular the mechanical loss factor, tan \(\delta\).

The fundamental method employed in this study was to use an acoustic speaker to supply a low frequency, sinusoidally varying acoustic stress to the skin. Simultaneously, specific points on the skin were illuminated with coherent laser light and the motions of the backscattered speckle patterns as a function of the driving acoustic stress wave were recorded. The magnitude and the time-resolved pattern of the speckle shift were calculated using the maximum likelihood estimator described above.

All of the experiments were performed using samples of fresh porcine flesh obtained from a local abattoir. The tissue samples consisted of skin, adipose tissue and underlying muscle tissue. Typical dimensions of the tissue sample were 20 cm X 10 cm X 5 cm in the x, y, and z dimensions, respectively. Tissues were kept refrigerated until approximately ½ hour before testing at which time they were removed from refrigeration, surrounded on all sides, except for the skin surface, with damp paper towels and allowed to equilibrate to room temperature. To apply the acoustic stress wave, a small speaker (87 W, 7.0 cm diameter) was placed in direct contact with the skin at one end of the sample. The speaker was sinusoidally driven at 1 Hz by a sine wave originating from a function generator. In addition, the speaker was assumed to have remained in solid contact with the skin during excitation. Optical illumination was supplied by a 5 mW green HeNe laser (\(\lambda = 543\) nm) coupled through single-mode fibers. This wavelength was chosen as it restricts our interrogation depth to depths on the order of 100 \(\mu m\) or so, thus the measurements are primarily surface measurements.

The measurement of tan \(\delta\) relies upon determining the relative phase difference between the driving acoustic sinusoid and the sinusoidal shift in the speckle pattern from a single illuminated spot. In this case, the illuminated spot was located 5 cm from the edge of the speaker. The speaker was again driven with a 1 Hz sinusoid from a function generator and the backscattered speckle pattern was imaged with the linear array CCD camera. As above, stacked speckle histories were generated and the speckle shift was plotted along with the driving waveform as a function of time (or record number).

Based on the measured Rayleigh wave velocity, \(C_{R}\), a correction was made for time of flight of the surface acoustic wave and the phase difference, \(\delta\), between the driving acoustic wave and the resulting speckle shift was calculated. The phase of each waveform was determined as the arctangent of the ratio between the imaginary and real parts of the waves. The tangent of the phase difference was taken to be the desired variable, tan \(\delta\). The phase delay due to the electronics (i.e., the time delay between the time the signal is sent to the speaker and the coil actually moves) was considered to be negligibly small and was ignored in this analysis. This procedure was conducted on 3 different locations each on 5 separate skin samples.

**Results**

Based on the phase differences between the driving acoustic sinusoid and the sinusoid describing the time-resolved speckle shift, the mean value of tan \(\delta\) was 0.14 ± 0.07, or a phase angle of approximately 8°. Table 1 shows the actual values for each
A phase angle of 8° physically means that the strain in the skin sample lagged behind the imposed acoustic stress by 8° when the skin was subjected to a dynamic loading regime at 1 Hz.

Table 1. Measured tan δ values for 5 skin samples. Each value is the mean of three measurements.

<table>
<thead>
<tr>
<th>Skin Sample</th>
<th>Tan δ (mean ± s.d.)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>0.166 ± 0.04</td>
</tr>
<tr>
<td>2</td>
<td>0.237 ± 0.12</td>
</tr>
<tr>
<td>3</td>
<td>0.154 ± 0.07</td>
</tr>
<tr>
<td>4</td>
<td>0.140 ± 0.04</td>
</tr>
<tr>
<td>5</td>
<td>0.057 ± 0.03</td>
</tr>
<tr>
<td>Overall</td>
<td>0.14 ± 0.07</td>
</tr>
</tbody>
</table>

Discussion

The variation in tan δ values between samples is consistent with the variation seen between individuals in Karzakov and Klochkov’s [11] study on the low frequency mechanical properties of skin tissues in the human arm. Potts et al. [12] also noted large variations between individuals. This variation simply reflects the natural variation in tissues, and the fact that the mechanical properties of skin are very dependent upon location on the body and physiological parameters. It has been shown, however, that the values of tan δ of skin can be varied in a consistent fashion through the local injection of cross-linking agents (decreases tan δ) or enzymatic agents to partially digest the tissue (increases tan δ) [13]. Clearly, the variation in the measured mechanical properties of skin makes it somewhat impractical to rely solely on absolute values of the mechanical properties for diagnostic purposes. However, it seems reasonable that relative hydration levels of the skin could be determined in this fashion. On the other hand, variations in the local tissue mechanical properties may provide a means to interrogate the tissue. For example, if a large area of skin, including a suspected pathological site, were to be illuminated and a sequence of speckle images were taken as the skin is acoustically stressed, then neighborhood variations in the speckle shift across the images could be determined by extending the speckle shift algorithm to 3 dimensions (the 3rd dimension being time). In this manner, the tissue surrounding the suspect tissue would serve as an internal 'control'. Stiffer tissue areas would exhibit less speckle motion, while less stiff areas would exhibit greater motion under the same acoustic stress, indicating local variations in mechanical properties.

References
