Stress Relaxation of Porcine Gluteus Muscle Subjected to Sudden Transverse Deformation as Related to Pressure Sore Modeling

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Computational studies of deep pressure sores (DPS) in skeletal muscles require information on viscoelastic constitutive behavior of muscles, particularly when muscles are loaded transversally as during bone-muscle interaction in sitting and lying immobilized patients. In this study, we measured transient shear moduli \(G(t)\) of fresh porcine muscles in vitro using the indentation method. We employed a custom-made pneumatic device that allowed rapid (2000 mm/s) 4 mm indentations. We tested 8 gluteus muscles, harvested from 5 adult pigs. Each muscle was indented transversally (perpendicularly to the direction of fibers) at 3 different sites, 7 times per site, to obtain nonpreconditioned (NPC) and preconditioned (PC) \(G(t)\) data. Short-term \(G_s\) and long-term \(G_l\) shear moduli were obtained directly from experiments. We further fitted measured \(G(t)\) data to a biexponential equation \(G(t) = G_1 \cdot \exp(-t/\tau_1) + G_2 \cdot \exp(-t/\tau_2) + G_{\infty}\), which provided good fit, visually and in terms of the correlation coefficients. Typically, plateau of the stress relaxation curves (defined as 10% difference from final \(G_{\infty}\) ) was evident \(\sim 20\) s after indentation. Short-term shear moduli \(G_s\) (mean NPC: 8509 Pa, PC: 5711 Pa) were greater than long-term moduli \(G_l\) (NPC: 609 Pa, PC: 807 Pa) by about an order of magnitude. Statistical analysis of parameters showed that only \(G_2\) was affected by preconditioning, while \(G_1, G_s, G_{\infty}, \tau_1, \tau_2\), and \(G_l\) properties were unaffected. Since DPS develop over time scales of minutes to hours, but most stress relaxation occurs within \(\sim 20\) s, the most relevant property for computational modeling is \(G_1\) (mean \(\sim 700\) Pa), which is, conveniently, unaffected by preconditioning. [DOI: 10.1115/1.2264395]

Keywords: pressure ulcer, decubitus, striated muscle, viscoelastic mechanical properties, indentation

1 Introduction

Deep pressure sores (DPS) are attributed to deficient blood perfusion and lymphatic drainage, to excessive local deformations in tissues, and to ischemia-reperfusion sequences. In immobilized (bedridden or wheelchair bound) patients, DPS are a life-threatening, costly, and common complication [1–4]. The sites most susceptible to DPS are the regions that make contact with the supporting surface and also contain bony prominences that tend to concentrate mechanical loads and stresses [1,5]. The majority of the wounds appear on the lower part of the body, mostly around the sacrum and ischial tuberosities (38–56%) [2,6].

Prolonged compression of vascularized soft tissues, which obstructs pathways of nutrient supply and waste clearance, is considered the most important mechanical cause for onset of DPS [1,5]. Prolonged excessive compression has been shown to cause necrosis of skin, subcutaneous tissues and striated muscle, but muscular tissue appears to be the most sensitive one [7]. When DPS are formed in muscle tissue, the clinical classification is typically of a severe injury that is expected to expand towards the skin [1,5,8]. Treatment is consequently difficult, and in most cases, imposes surgery to remove necrotic tissues [5]. Biomechanical models showed that mechanical properties of the muscle play a dominant role in the prognosis of DPS. Specifically, exposure of striated muscle tissue to intensive and prolonged compression may pathologically alter its microstructure, thereby causing a change in tissue stiffness [1,8]. Changes in the mechanical properties of injured muscle tissue affect the mechanical stress and strain distributions around the site of the injury, and may expose additional uninjured regions of muscle tissue to intensified stresses in a positive-feedback mechanism that widens the injured area [1,8]. Therefore, understanding the etiology of DPS requires information on internal stress and strain distributions in deep tissues, particularly in the muscles underlying bony prominences, although that is less clear for the gluteus muscles which envelop the ischial tuberosities in these postures [1,5,8].

Biomechanical models employed for this purpose make use of constitutive (stress-strain) relationships to describe tissue behavior. However, a problem in formulating these constitutive equations is the limited availability of experimental data on tissue mechanical properties. Specifically, there are very limited published data on viscoelastic mechanical properties of striated muscle under transverse compression [8–10], i.e., compression of muscle tissue perpendicularly to the direction of muscle fibers, as in the case of DPS biomechanics, where bony prominences (e.g., the sacrum and ischial tuberosities) laterally compress the muscles (e.g., the longissimus and gluteus, respectively).

Published data of transverse viscoelastic mechanical properties for skeletal muscles, obtained in vitro [9] and in vivo [8,10], are all from rats, and so, there is paucity of information regarding transverse muscle properties in larger animal models, particularly in the pig, which is commonly used in pressure sore studies [7,11]. Another problem is that previous protocols of muscle transverse loading [8,10] included ramp-and-hold tests with relatively long transition periods (of 1–10 s, depending on the size of the step deformation), rather than a (nearly) instantaneous step deformation. The response to a rapid step deformation is required to accurately evaluate the viscoelastic response (particularly in short terms) to any general wave shape of deformation in absence of a specific constitutive model, based on Boltzmann’s principle of superposition in linear viscoelasticity, or based on Fung’s quasilinear viscoelastic theory [12].

In this study, rapid indentation experiments were conducted using a custom-made pneumatic stress-relaxation testing device that produced minimal transition periods (as opposed to slow step-motor-driven systems [8,10]), thereby measuring more accurate relaxation time courses, particularly in the short-term phase of tissue response. Determining the time course of relaxation of muscle tissue in the transverse direction is significant for DPS biomechanics because recently, it was indicated that pressure sores can develop in muscles on a time scale of minutes-to-hours, not necessarily hours-to-days, as it has long been believed [8]. For biomechanical modeling of the development of DPS, it is required to know when the mechanical properties of muscle tissue can be considered stable (after viscoelastic stress relaxation), with respect to the time when a patient was put in the immobilized position.
that led to DPS. If significant stress relaxation still continues when DPS onset/progress, this must be considered in the modeling [1,5,8]. Nonetheless, we hypothesize that the long-term viscoelastic characteristics (e.g., the long-term shear modulus) are the ones relevant to DPS modeling. Accordingly, the specific aims of this study were (i) experimental characterization of the in vitro shear moduli of fresh porcine gluteus muscle and characterization of the time course of relaxation in response to a (nearly ideal) step deformation in the transverse direction, and (ii) mathematical formulation of this response, in a form that can be used for future analytical and computational models.

2 Materials and Methods

2.1 Experimental Apparatus and Protocol. Muscle tissue is a complex hierarchical structure. At a microscopic scale, muscle tissue contains muscle fibers packed together by endomysium connective tissue, and individual packs are wrapped by perimysium connective tissue. Although this heterogeneous structure generally exhibits anisotropic mechanical behavior, for the purpose of biomechanical measurements at the macroscopic level, it is common to approximate muscle tissue as being a homogenous isotropic material [1,8]. This assumption allows a transient shear modulus of muscle tissue during viscoelastic stress relaxation $G(t)$ to be calculated, if the muscle is also considered a semi-infinite volume that is indented by a rigid indenter with a circular flat tip [8,13].

$$G(t)/(1 - \nu) = [F(t)/\delta]/(2\pi d)$$

where $\nu$ is the Poisson’s ratio, $d$ is the diameter of the indenter (taken here is 12 mm), $\delta$ is the depth of indentation (set as 4 mm), and $F(t)$ is the measured time-dependent relaxation force. For viscoelastic materials such as muscle tissue, the load $F(t)$ decreases with time to an asymptote during the “hold” period of the indentation. Accordingly, for a step indentation, a short-term (instantaneous) shear modulus $G_s$ can be calculated from Eq. (1) using the peak, or instantaneous load measured immediately at the start of the “hold” period. A long-term (asymmetric) shear modulus $G_l$ can be calculated using the asymptotic load value. Equation (1) was successfully used in previous studies to determine the transient shear modulus $G(t)$ and its characteristics $G_s$ and $G_l$ for many soft tissues [13]. In the present study, we employed Eq. (1) to determine the transient shear modulus $G(t)$ of fresh porcine skeletal muscle tissue in the transverse direction (i.e., perpendicular to the direction of muscle fibers). In order to solve Eq. (1) for the transient shear modulus $G(t)$ of skeletal muscles, we set $\nu = 0.5$, considering that muscle tissue contains approximately 75% water and is therefore nearly incompressible [8].

Porcine gluteus muscle was selected for this study due to anatomical and physiological similarities to human skeletal muscles [14] and because the pig is a well-accepted model for human skeletal muscles. The muscles were kept moist during experiments using a saline spray. The muscles were immersed in saline just after dissection, and were transported at 4°C to the testing facility, where recording of the relaxation responses began within 20 min from dissection, and was completed within 3 h from the time of euthanasia, to ensure that no rigor mortis occurred [17,18]. Muscles were tested in air, after being allowed a few minutes to equilibrate with room temperature, so that muscle temperature was ~24°C during measurements. The muscles were kept moist during experiments using a saline spray.

In vitro indentation experiments were conducted using a custom-made pneumatic testing device (Fig. 1) that has a swift indentation speed of 2000 mm/s during the ramp phase (compared with an indentation speed of 0.25–1 mm/s in previous studies [8,10]). Hence, the ramp phase of 4 mm indentations (4 mm was set as the constant indentation depth in all experiments) lasted 0.002 s. Such swift indentations may induce transient stress waves in the tested specimens. To verify that transient stress waves decayed before influencing measurements of long-term shear moduli—the property hypothesized herein to be the most relevant for pressure sore modeling—we conducted experiments with a viscoelastic polyurethane gel specimen. The size, thickness, and stiffness (long-term shear modulus ~1000 Pa) of the gel specimens were similar to those of our muscle specimens. Use of a synthetic phantom, rather than tissue, eliminated the effect of biological variability and allowed isolation of the stress wave effect on long-term shear moduli $G_l$. We conducted 10 trials of swift indentations (0.002 s risetime of ramp), and 10 additional trials in which we loaded the viscoelastic gel quasi-statically (which ensured that no stress waves occurred). In all swift indentation tests, stress waves appeared in the polyurethane gel specimen within the first 0.01 s of relaxation, with magnitude, frequency, and decay patterns that were similar to those seen during muscle tissue indentations. However, the long-term shear moduli $G_l$ obtained in the quasi-static and swift indentation trials were statistically indistinguishable ($p$-value of 0.85 in an unpaired two-tailed t-test). We concluded that transient stress waves did not affect the long-term shear moduli $G_l$ in our experimental setup, and, hence, the same deformation parameters (0.002 s risetime of ramp deformation and 4 mm depth of indentation) were used for testing muscle specimens.

Each muscle was indented 7 times at 3 different sites that were at least 5 cm apart on the surface of the muscle’s central third, in order to obtain both nonpreconditioned (NPC) and preconditioned (PC) viscoelastic mechanical properties. We verified that the rapid indentations driven by the pneumatic device did not cause tears or any other type of visible damage to the muscle surface. At each indentation site (3 sites per muscle), NPC results were obtained from the first indentation run, and PC results were obtained from the last 3 indentation cycles. The last 3 indentations at each site were assigned as PC data because in preliminary muscle tissue testing as well as for the muscles analyzed herein, stress relaxation curves overlapped visually and were shown to be stable for the last 3 indentation cycles (out of 7) by two separate one-way analysis of variance (ANOVA) tests comparing the fifth, sixth, and seventh cycles of $G_s$ and $G_l$. These tests showed that both $G_s$ and $G_l$ properties were statistically indistinguishable across the last 3 preconditioning cycles, indicating arrival at a stable stress relaxation response.

The duration of each measurement and the pause between subsequent preconditioning indentations at the same site (which allowed elastic recovery of tissue) was set as 60 s, due to experimental design limitations, which required that all measurements be completed early enough before rigor mortis may onset. However, the relaxation response was observed to plateau well within the 60 s duration. Specifically, $G(t)$ reached 10% difference from the final value ($G_f$) after a mean time of ~20 s, and 2–4% difference from the final value ($G_f$) after a mean time of ~40 s.

In each stress relaxation trial, the indenter, with a circular, flat tip (diameter 12 mm), was placed over the central part of the muscle and carefully lowered to the surface using its system of axial adjustors and elevator (Fig. 1). As soon as indenter-muscle contact was detected, as determined by a threshold reading of 0.1 N from the force transducer attached to the indenter, position was held and the pneumatic piston (BDALS 10X30, Koganei Co., Japan) driving the indenter was armed at a constant air pressure of 400 kPa as measured by a pressure regulator (to ensure identical conditions across all measurements). A stopper plate was set at a distance of 4 mm from the surface of the indenter to ensure that...
all indentations are to the same 4 mm depth. After release of the pressurized pneumatic piston, force readings were recorded over time for each indentation run using a force transducer attached to the indenter. The force transducer has a response time of less than 5 μs, capacity of 20 N, accuracy of ±2.5%, and drift <3% per logarithmic time scale (Flexiforce A201, Tekscan Co., MA, USA). Force signals were amplified using operational amplifier (LM324N, National Instruments Co., TX, USA) and were sampled continuously at a frequency of 1000 Hz for the first 3 s and at 100 Hz afterwards (using A/D card and LabView 7.0 software, National Instruments Co.) to maintain reasonable data file sizes (Fig. 2). Results from two of the muscles and ~10% of stress relaxation trials from the other 8 muscles were excluded due to interrupted indenter contact manifested by noisy data.

2.2 Data Analysis. Since DPS develop over time scales of minutes-to-hours [8], but most stress relaxation was shown to occur within ~20 s (as indicated above), the most relevant property for DPS modeling studies emerges to be the long-term shear modulus of muscles $G_L$. Considering that once an immobilized person was positioned in a bed or wheelchair, he is not expected to change posture for a long period of time, the NPC modulus, particularly, seems to be relevant for computational modeling of DPS. Nevertheless, we acquired short-term and long-term, NPC and PC shear moduli for completeness. Hence, short-term (instantaneous) shear moduli ($G_S$) and long-term shear moduli ($G_L$), were obtained directly from substituting the experimental force data in Eq. (1). Specifically, we calculated $G_L$ from each stress relaxation trial by substituting the peak force, max(F(t)), measured at the instance of deformation. The corresponding $G_L$ was calculated by substituting the mean of force data over the terminal 1 s (100 data points) of the plateau phase of the stress relaxation curve (we defined the beginning of the plateau phase as the time point at which $G_L$ reached 10% difference from the final value). We then calculated means and standard deviations of experimentally obtained, NPC and PC $G_S$ and $G_L$ values, as follows. First, to reduce prospective measurement errors in PC $G_S$ and $G_L$ data, and considering that ANOVA showed stable, repeatable stress relaxations during the last 3 preconditioning cycles (both $G_S$ and $G_L$ were statistically indistinguishable across the fifth, sixth, and seventh preconditioning cycles), we averaged each parameter across the 3 last indentation runs at each site, and termed the mean values from the 3 last runs as the fully PC $G_S$ and $G_L$ data. Second, in order to further reduce variability, we averaged the NPC and fully PC $G_S$ and $G_L$ data (separately) across the 3 indentation sites in each muscle, and hence, obtained the effective NPC and PC $G_S$ and $G_L$ properties per muscle.

Using the Matlab (version 7) software “curve fitting tool,” the experimental transient shear moduli $G(t)$ were found to have a good fit to a biexponential equation [19], both visually and in terms of correlation coefficients (for a total of 43 NPC and fully PC curves: $R^2>0.9$ for 37 curves, $R^2>0.85$ for 5 curves, and $R^2>0.7$ for one curve). Experimental $G(t)$ data were therefore fitted to a function given by

$$G(t) = G_1 \cdot \exp(-t/\tau_1) + G_2 \cdot \exp(-t/\tau_2) + G_0$$

where $\tau_1$ and $\tau_2$ are the short and long relaxation times, $G_1, G_2$ are constants, and $G_0$ is the long-term shear modulus from curve fitting. Substituting $t=0$ in Eq. (2) yields that the short-term (or
The values of $G_1, G_2, G_{\infty}, \tau_1, \tau_2$, the parameters of the transient relaxation modulus function (Eq. (2)), were obtained for the naive NPC indentation and the 3 last indentation runs at each site (Fig. 2). Similarly to the above-described processing of directly measured $G_S$ and $G_T$ values, in order to obtain mean PC parameter values from curve fitting, we averaged $G_1, G_2, G_{\infty}, \tau_1, \tau_2$ parameters across the 3 last indentation runs at each site, and referred to the mean values from these last 3 maneuvers as the fully PC data. In order to further reduce variability in $G_S$, we averaged each parameter across the 3 sites per muscle specimen (separately for the NPC and PC results), so that an effective mechanical characterization of the NPC and PC relaxation response and transient shear modulus $G(t)$ per muscle was obtained.

2.3 Statistical Testing. All data sets were tested for normal distribution using the Shapiro-Wilk test, and all parameter values were found to be normally distributed with the exception of PC values of $\tau_2$. However, PC $\tau_1$ values were distributed very close to a normal distribution as was evident from the linearity of a normal quantile plot drew for this parameter. Statistical outlier values were removed from the $G_S, G_{L}, G_1, G_2, G_{\infty}, \tau_1, \tau_2$ data sets, based on the results of a Grubb’s test run separately for each parameter [20]. We typically removed 1–3 outlier values per parameter. Paired t-tests were then conducted for NPC versus PC values of measured short-term ($G_S$) and long-term ($G_T$) shear moduli across all muscle specimens to determine if these shear moduli are affected by preconditioning. “One-tail” tests were employed since preconditioning a soft tissue in vitro or in situ was shown to consistently decrease the short-term and long-term shear moduli at the preconditioning site [8,19]. In all statistical tests, a $p$-value lower than 0.05 was considered statistically significant.

3 Results

Table 1 shows the mean values and standard deviations obtained for all parameters, $G_S, G_L, G_1, G_2, G_{\infty}, \tau_1, \tau_2$, during NPC and PC indentations. Our measured variations in mechanical properties, due to biological variability, are similar to those reported in previous studies [9,17,19]. Short-term shear moduli $G_S$ were in the order of thousands of Pascals, and long-term shear moduli $G_L$ were in the order of hundreds of Pascals (Table 1). Hence, stress relaxation was shown to decrease NPC shear moduli 14-fold, and likewise, decrease PC shear moduli sevenfold (Table 1). We studied the effect of preconditioning on measured short-term shear moduli ($G_S$) and long-term shear moduli ($G_L$) of porcine gluteus muscles by means of paired t-tests. Preconditioning was shown to significantly decrease mean $G_S$ (~0.67-fold, Table 1) ($p=0.04$). However, we found that the effect of preconditioning on $G_L$ was not statistically significant ($p=0.32$). It should be noted that although mean PC $G_L$ values appear to be higher than mean NPC $G_L$ (~1.3-fold, Table 1), individual sites in a muscle always displayed a decrease of $G_L$ values with preconditioning (though not a statistically significant decrease), and so, the difference between means of NPC and PC $G_L$ should be attributed to biological variability and not to preconditioning. A retrospective power analysis indicated that the chance of detecting a statistically significant effect of preconditioning on $G_L$ with our sample size was 62%, and that the minimal detectable difference in testing NPC versus PC $G_L$ was 298 Pa (for “one side” test with power of 90%). Accordingly, it is possible that preconditioning might have affected $G_L$ to an extent smaller than our above test resolution. Taking this reservation into account, we conclude that $G_L$, aforementioned as the most relevant property for DPS related computational modeling, is conveniently found to be unaffected by preconditioning.
Table 1 Parameter values for porcine gluteus transient shear modulus of the form: $G = G_0 \cdot \exp(-t/\tau) + G_s$, and experimental values for short-term and long-term shear moduli.

<table>
<thead>
<tr>
<th>Parameter</th>
<th>NPC</th>
<th>PC</th>
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<tr>
<td>$G_0$ (Pa)</td>
<td>466 ± 3775</td>
<td>3922 ± 2520</td>
</tr>
<tr>
<td>$G_s$ (Pa)</td>
<td>249 ± 157</td>
<td>3922 ± 350</td>
</tr>
<tr>
<td>$\tau$ (s)</td>
<td>0.05 ± 0.035</td>
<td>0.006 ± 0.019</td>
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The biexponential curve fitting (Eq. 2) provided excellent approximations of the long-term shear modulus $G_L$ with respect to measured $G_L$ (error of 6% and 0.5% for the NPC and PC values, respectively, Table 1). However, errors in approximating the short-term modulus $G_S$ by $G_0 + G_s$ were more substantial (error of 25% and 11% for the NPC and PC values, respectively, Table 1). As already noted, stress relaxation was shown to occur rather quickly, and hence, 90% of the full relaxation ($G_L$) was reached within ~20 s (mean) from the instant of deformation (Fig. 2). Relaxation times were more than two orders of magnitude apart, with the short relaxation time constant being less than 0.1 s, and the long time constant being less than 20 s (Table 1). We further notice that the ramp risetime induced by the pneumatic testing device (0.002 s) is substantially shorter than the short relaxation time constant $\tau_1$ (~26-fold NPC, ~eightfold PC, Table 1), which ensures that information on the short-term phase of stress relaxation was adequately captured [12].

4 Discussion

The objective of the present study was to characterize the viscoelastic stress relaxation and shear moduli of porcine gluteus muscles in vitro, in a manner suitable for analytical and computational studies of DPS in humans. Such modeling generally requires that the viscoelastic mechanical properties of muscles be specified, particularly, in the transverse direction (perpendicularly to muscle fibers) to mimic compression of bony prominences in sitting or lying humans. The results provided herein can be employed to describe the time course of stress relaxation of glutus muscles in the transverse direction, in full detail. In many DPS related studies, however, which deal with prolonged excessive compression of muscle tissue by bony prominences in immobilized patients [1, 5, 8], it is mainly the long-term shear modulus $G_L$ (or $G_S$) that is of interest. The present results indicate that stress relaxation in the transverse direction of muscle tissue occurs within less than 1 min, i.e., shorter than the minutes-to-hours time scale required for developing DPS [8]. Hence, $G_L$ in the transverse direction is indeed the relevant mechanical property when modeling muscle tissue as related to DPS.

Previous studies showed that mechanical properties of muscle tissue reach a steady state response, and therefore, become repeatable as a result of preconditioning [8, 21]. It has been shown herein that for the sample size considered in this study, $G_L$ was statistically unaffected by preconditioning, and this may simplify DPS modeling, particularly when it is desirable to take postural changes of patients (which induce muscle preconditioning) into account. It should be noted that the exclusion of statistical outliers, averaging of data, and selection of a one-tailed statistical analysis considerably increased the probability that differences resulting from preconditioning would be detected. However, in spite of that, the effect of preconditioning on long-term moduli was not significant, further supporting the main result of this study. The latter finding, that $G_L$ is not affected by preconditioning, is also in accordance with a previous study of soft tissue viscoelasticity, which found that preconditioning does not affect the long-time behavior of human ankle ligament [22]. Taken together, our findings therefore indicate that $G_L$ of skeletal muscles in the transverse direction should be taken as ~700 ± 300 Pa (NPC and PC combined). This value is in exact agreement with in vivo $G_L$ data (for transverse loading) which were previously obtained from gracilis of anesthetized rats [8], and with a literature review suggesting that the long-term shear moduli of human skeletal muscle in the transverse direction should be in the order of 250–1200 Pa [8].

This study had some limitations which should be considered while interpreting the results. The first limitation concerns the assumption that muscle tissue is homogenous. This assumption allowed characterization of an effective viscoelastic mechanical behavior at the continuum level, but not local variants of tissue.
mechanical properties. However, the effective viscoelastic properties at the continuum level of muscle are sufficient for most DPS modeling tasks [1,8]. The analytical solution of the indentation problem (Eq. (1)) further assumes homogenous elastic tissue with a perfectly flat surface and boundaries at an infinite distance from the site of indentation. The latter assumption was closely approached because the indenter’s diameter was small compared to the distance between the indenter and the tissue boundaries. A second limitation is that the $G(t)$ experimental data presented here are subject to a time limit of 60 s, the “hold” time of the stress relaxation tests. This time limit was necessary, because experiments had to be carefully limited in time frame to ensure that measurements are taken prior to rigor mortis effects [17,18]. Determination of 60 s data capture periods may result in some small overestimation of the viscoelastic shear modulus, due to approximation of the point at which the stress relaxation test is ended as an infinite time [8,12,22]. Considering, for example, the case of a wheelchair bound patient whose posture is changed once every 30 min [8]. The mean rate of change in (PC) shear moduli at the terminal second of measurement ($\frac{dG_s(t)}{dt} = 0.05 \text{ Pa/s}$, and hence, further decrease in $G_t$ due to ongoing stress relaxation of muscles in such patient who is immobilized for 30 min (1800 s) must be smaller than $-0.05 \text{ Pa/s} \times 1800 \text{ s} = -90 \text{ Pa}$, i.e., maximal $\sim 13\%$ decrease from the presently measured $\sim 700 \text{ Pa}$ value. Considering also the biological variability in $G_t$ properties (ratio of standard deviation over mean $\sim 35\%$ for NPC and $\sim 50\%$ for PC $G_t$ properties), long-term shear moduli obtained after a longer relaxation period are expected to be similar to those obtained herein. A third limitation relates to the rapid rate of indentations. We employed swift indentions (0.002 risetime of ramp deformation) to obtain a relaxation response that is as similar as possible to the theoretical “step response.” However, as opposed to theoretical (ideal) step deformations, swift deformations in an experimental system may involve stress waves, some of which are apparent in the magnification of the short-term response in Fig. 2 (top right frame). These stress waves were shown to damp out completely within less than 0.5 s from the onset of deformation. Nonetheless, we verified, using phantom experiments, that $G_t$, the relevant property for DPS modeling, is unaffected by these stress waves (see Methods). Unfortunately, we cannot exclude a possibility that the short-term transient constants of the relaxation response $\tau_1$ and $G_t$, which depend on the deformation rate [12], might have been affected by stress waves. Last, the viscoelastic properties reported herein represent passive muscle behavior, and do not consider stiffness due to muscle contractility. However, as this study is relevant to modeling the mechanical conditions in muscles of immobilized (wheelchair bound and bedridden) patients who typically have no neural stimulation of muscles, the passive properties are of interest.

In closing, further work in applying various analytical and computational models to the study of DPS can make use of the quantitative formulation of the stress relaxation and shear moduli of porcine gluteal muscle tissue in the transverse direction, which are reported in this study.

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