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Office-Based Elastographic Technique for Quantifying Mechanical Properties of Skeletal Muscle

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Abstract

Objectives—Our objectives were to develop a new, efficient, and easy-to-administer approach to ultrasound elastography and assess its ability to provide quantitative characterization of viscoelastic properties of skeletal muscle in an outpatient clinical environment. We sought to show its validity and clinical utility in assessing myofascial trigger points, which are associated with myofascial pain syndrome.

Methods—Ultrasound imaging was performed while the muscle was externally vibrated at frequencies in the range of 60 to 200 Hz using a handheld vibrator. The spatial gradient of the vibration phase yielded the shear wave speed, which is related to the viscoelastic properties of tissue. The method was validated using a calibrated experimental phantom, the biceps brachii muscle in healthy volunteers ($n = 6$), and the upper trapezius muscle in symptomatic patients with axial neck pain ($n = 13$) and asymptomatic (pain-free) control participants ($n = 9$).

Results—Using the experimental phantom, our method was able to quantitatively measure the shear moduli with error rates of less than 20%. The mean shear modulus $\pm$ SD in the normal biceps brachii measured 12.5 ± 3.4 kPa, within the range of published values using more sophisticated methods. Shear wave speeds in active myofascial trigger points and the surrounding muscle tissue were significantly higher than those in normal tissue at high frequency excitations (>100 Hz; $P < .05$).

Conclusions—Off-the-shelf office-based equipment can be used to quantitatively characterize skeletal muscle viscoelastic properties with estimates comparable to those using more sophisticated methods. Our preliminary results using this method indicate that patients with spontaneous neck pain and symptomatic myofascial trigger points have increased tissue heterogeneity at the trigger point site and the surrounding muscle tissue.

Keywords

biomechanics; myofascial trigger points; shear wave elastography; ultrasound

There is a clinical need to quantitatively evaluate the mechanical properties of skeletal muscle in vivo.^{1,2} Until recently, only in vitro methods were available for quantifying mechanical properties of tissue. In the past decade, new in vivo imaging techniques have

been developed to quantitatively estimate shear mechanical properties of soft tissue, including skeletal muscle, via magnetic resonance elastography^{3,4} and ultrasound (US) elastography.^{5,6} These quantitative elastographic methods attempt to estimate the speed of a propagating shear wave in soft tissue, which depends on the underlying mechanical properties. The shear wave may be induced externally⁷ or may use the radiation force of US.⁸ For viscoelastic materials, such as most soft tissue, the mechanical properties include not only the elastic moduli but also the viscous moduli. If the dispersion of the shear wave speed with the frequency can be estimated, the Voigt model can be used to estimate both the elastic and viscous moduli.^{9,10} This approach has been previously used to measure shear properties of the liver,^{10,11} breast,^{12,13} bicep,^{5,10} trapezius,^{3,4} and rectus femoris muscle.¹⁴ However, these proposed methods have traditionally relied on specialized equipment.

Our objective was to develop and evaluate a method for quantitative elastography of skeletal muscle based on a clinical US system and off-the-shelf components. We validated our elastographic technique by quantifying shear properties of an experimental phantom as well as in vivo biceps brachii muscle, which has been well studied in the literature.^{5,10,15} To further show the clinical utility of our method, we investigated the viscoelastic properties of the upper trapezius muscle in healthy participants as well as patients with neck pain. The upper trapezius is of great interest because it is frequently painful and the site of the commonly occurring myofascial trigger points. These trigger points are stiff, discrete, palpable, tender nodules that can be found in the belly of a taut band of muscle and are a characteristic finding in myofascial pain syndrome, which is thought to affect almost 23.8 million Americans each year.¹⁶

Myofascial pain syndrome is a descriptive term used to define an acute or chronic soft tissue musculoskeletal pain condition that is characterized by sensory, motor, and autonomic findings associated with myofascial trigger points.¹⁷ These trigger points can exist in active or latent states. Active myofascial trigger points are associated with spontaneous pain, are acutely tender to palpation, and may contribute to general motor dysfunction (stiffness and a restricted range of motion). Latent myofascial trigger points, which have similar physical findings, are often associated with motor dysfunction and muscle tenderness but are only painful on firm palpation. Although not spontaneously painful, latent myofascial trigger points are very common clinical findings in both symptomatic and asymptomatic individuals.^{18,19} Therefore, in someone with spontaneous pain, thorough palpation of the myofascial tissue is required to clinically identify and differentiate an active from a latent myofascial trigger point. Normal or uninvolved muscle does not contain taut bands or myofascial trigger points.

The current diagnostic standard for myofascial pain is based on palpation for the presence of trigger points in a taut band of skeletal muscle and an associated symptom cluster that includes referred pain patterns.²⁰ Unfortunately, few physicians receive training in the clinical diagnosis of myofascial pain. Moreover, the physical examination has been reported to be unreliable and is not sensitive enough to changes in the soft tissue properties.²¹ Until recently, there have been no clearly demonstrable underlying pathologic characteristics associated with the physical findings of trigger points and taut bands.^{3,17,22–24} However, the physical exam remains the gold standard for diagnosis of myofascial trigger points. In our research group, we are developing objective image-based measures to overcome these limitations. We have previously shown that myofascial trigger points can be imaged using US.^{25,26} In an effort to further develop quantitative assessments of trigger points and their surrounding soft tissue milieu, in this study we used our elastographic method to study participants who were either “normal” (ie, pain free and having no palpable myofascial trigger points) or “active” (having 1 or more palpable trigger points with spontaneous pain in the upper trapezius). We hypothesized that the soft tissue mechanical properties of active

myofascial trigger points and the surrounding muscle tissue around the trigger points in symptomatic patients are quantitatively different from the normal muscle tissue in asymptomatic (ie, pain-free) individuals, thus potentially enabling the development of quantitative outcome measures in the future.

Materials and Methods

Shear Wave Elastography Using a Handheld Vibrator

A handheld vibrator was fabricated using a direct current motor with an offset weight attached to its rotor axis (7.8V mini direct current motor; Battery Space, Richmond, CA) contained in a plastic housing (Figure 1). A variable power supply (0–30 VDC Lab power supply; All Electronic Corp, Van Nuys, CA) was used to drive the motor at approximately 4 different frequencies: 60, 110, 160, and 200 Hz. The actual frequency of the vibration was measured using US, as described below.

Ultrasound imaging was performed using a Sonix RP US system (Ultrasonix Corp, Vancouver, British Columbia, Canada) and a 5- to 14-MHz linear array transducer. The raw radiofrequency (RF) data were collected at a high frame rate of 522 frames per second. Each frame consisted of 64 scan lines, a 50% sector size, and an imaging depth of 2.5 cm. The data were processed online using an Uterius software development kit (Ultrasonix Corp) and MATLAB (The MathWorks, Natick MA) to produce shear wave phase angle images as well as B-mode images.

Because data acquisition was very sensitive to the positions of the US transducer and the vibration generator, a preview function was generated (in MATLAB, using data streaming from the Uterius program) that ran in parallel and produced the phase images immediately after the data capture. The interactive phase images were used to verify that the propagating shear wave was captured and to ensure that both the vibration probe and the US imaging probe were positioned so that they were aligned along the muscle fibers. The data acquisition was repeated if a propagating shear wave was not satisfactorily visualized. This process ensured repeatability of the method and minimized operator dependency.

Signal Processing to Estimate the Shear Wave Speed

The received RF data were processed using a Hilbert transform to generate the complex analytical signals and were demodulated to baseband. The envelope of the analytical signals formed the B-mode image. A conventional autocorrelation algorithm was used to estimate the instantaneous speed of the tissue.²⁷ The instantaneous speed was filtered using a 2-dimensional median filter (0.125 mm axial and 1.5 mm lateral). A moving average filter of length 0.25 mm in the axial direction was used to suppress isolated phase peaks at speckle boundaries. The temporal frequency of the shear wave was estimated using the Fourier transform of the filtered instantaneous speed after a 0.25-second Hanning window was applied and the mean was removed. The excitation frequency was calculated as the mode of the magnitude of the maximum frequency peak for each axial/lateral position. The phase of the Fourier transform at this excitation frequency yielded the phase of the shear wave (Figures 2 and 3). A correction was made for the phase delay due to the difference in the timing of the US transmission for different scan lines as follows: $\theta_k^* = \theta_k \pm k\omega_s\Delta t$, where θ_k was the uncorrected phase in radians at the k th scan line; ω_s was the angular frequency of the excitation in radians per second; Δt was the round-trip time in seconds for a single scan line; and the positive or negative sign is chosen depending on whether the external vibrator is placed closest to the first or the last scan line. This phase adjustment was made before computing the shear wave speed.

A region of interest was then selected on these phase images, and the shear wave propagation speed, C_s , was estimated from the spatial gradient of the phase using

$$C_s = \omega_s \frac{\Delta r}{\Delta \theta},$$

where $\Delta \theta$ was the difference in the phase angle over the lateral distance (Δr), and ω_s was the angular shear wave vibration frequency in radians (Figures 2 and 3). For each region of interest, an average phase gradient was estimated from 3 measurements. These shear wave propagation speeds at different frequencies were then fit to a Voigt model for viscoelastic materials to estimate the shear modulus (μ_1) and shear viscosity (μ_2) as follows:

$$C_s = \sqrt{\frac{2(\mu_1^2 + \omega_s^2 \mu_2^2)}{\rho(\mu_1 + \sqrt{\mu_1^2 + \omega_s^2 \mu_2^2})}},$$

where ρ was the tissue density (1060 kg/m^3). The values for the shear modulus and shear viscosity were determined using a least squares fit of measured shear wave velocities to the Voigt model using the generalized reduced gradient algorithm (standard optimization algorithm implemented in the Excel Solver tool; Microsoft Corporation, Redmond, WA). The spatial resolution of our current analysis method for shear wave speed measurement was approximately 4 to 5 mm for a given vibration frequency, including all of the filtering and slope estimations. The size range of myofascial trigger point nodules is 0.2 to 2.0 cm^{2,26} so the spatial resolution was adequate for imaging the trigger points.

In Vitro Validation Using an Experimental Phantom

To validate the ability of our proposed technique to detect localized differences in mechanical properties, we used a hydrogel-based US tissue-equivalent phantom (049 Elasticity QA phantom; CIRS Inc, Norfolk, VA), which consisted of a background material and 4 types of spherical inclusions of different elastic properties. Shear wave elastography was performed on the background alone (shear modulus $G = 8.2 \text{ kPa}$), the background with a stiffer type IV sphere ($G = 24.9 \text{ kPa}$), and the background with a less stiff type I sphere ($G = 2.9 \text{ kPa}$). The shear modulus and shear viscosity were measured for all material types, including the background material where spheres were present, to ensure repeatability of the background material property estimates even in the presence of stiffer or less stiff material.

In Vivo Validation Using the Biceps Brachii

The viscoelastic properties of the biceps brachii have been extensively studied in the literature. To validate our proposed technique in vivo, we estimated the shear modulus and shear viscosity properties of the biceps brachii in a group of healthy volunteers ($n = 6$) for comparison to literature values. The Human Subjects Review Board of George Mason University approved all of the procedures. The participants were seated comfortably, with their dominant arm resting on a table with their elbow bent at 90° and instructed to relax their arm and not flex/contract their biceps muscles. The long head of the biceps brachii was imaged using US while simultaneously being vibrated with the handheld vibrator at 4 different frequencies.

Clinical Validation in Patients With Palpable Myofascial Trigger Points

To evaluate the clinical applicability of our proposed technique, we performed a preliminary study of the biomechanical properties of the upper trapezius muscle, comparing patients with axial neck pain and palpable myofascial trigger points to a group of asymptomatic healthy volunteers with palpably normal muscle. We compared our elastographic findings to the findings of palpable trigger points on a physical examination.

Patient Selection and Physical Examination

Men and women with cervical pain who met inclusion criteria (ie, having an active myofascial trigger point in one or both upper trapezii) underwent a thorough musculoskeletal evaluation to rule out potential causes of their symptoms other than myofascial trigger points. Patients were classified as chronic if they had pain consistently for more than 3 months or acute if their pain duration was less than 3 months. Patients with posttraumatic pain were also included in the study unless they met the exclusion criteria, which included fibromyalgia, atypical facial neuralgia, myopathy, radiculopathy, history of shoulder or spine surgery, or previous trigger point injections.²⁵ A group of healthy volunteers with no neck pain were also recruited. A total of 22 participants were studied: 9 healthy participants (ie, no palpable myofascial trigger points) and 13 active patients (ie, those with palpable active trigger points). The mean ages $\pm$ SD and male to female ratios for the normal and active participants were 30 ± 5 years with 6 men and 3 women and 41 ± 13 years with 5 men and 8 women, respectively. Each participant provided informed consent to participate in the study. The study was performed at two sites: National Institutes of Health and George Mason University. Patients with acute neck pain (<3 months' duration; $n = 7$) were recruited at the National Institutes of Health, whereas patients with chronic neck pain (>3 months' duration; $n = 6$) were recruited at George Mason University. The study procedures were approved by the Institutional Review Boards of the National Institute of Dental and Craniofacial Research and George Mason University.

The participants underwent a physical examination as previously described.²⁵ Briefly, the presence or absence of myofascial trigger points in the upper trapezius muscle was determined by the criteria of Travell and Simons²⁰ according to standard clinical practice. Palpation was performed in the central region of the upper trapezius muscle within 6 cm of the muscle's midline (approximately midway between the cervical vertebrae and the acromion process). The 6-cm range was chosen as an approximate guideline to avoid the muscle attachment sites and stay in the central region of the muscle. Sites were normal if no palpable nodule was found. Active sites had at least 1 palpable nodule that was both spontaneously painful and caused exacerbation of the participant's symptomatic (not all had referred) pain on palpation. For participants with more than 1 active myofascial trigger point, the most symptomatic trigger point was selected for analysis. Measurements were repeated twice for each frequency. Only the examiner knew the clinical status of the participants (ie, whether they had cervical pain) and classifications of the marked sites. The sonographers were blinded to the clinical status when acquiring US data. Approximately 30 minutes elapsed between the physical and US examinations; the entire test procedures were conducted in less than 2 hours.

Typically, palpable myofascial trigger points appear as focal hypoechoic (darker) areas with a heterogeneous echo texture on gray-scale sonograms as previously described.²⁵ In this study, hypoechoic areas were first detected using B-mode imaging. Then an external vibration, at around 100 Hz, was applied to the surrounding muscle, and a color Doppler variance image was acquired. Myofascial trigger points, being stiffer, vibrate with a lower amplitude and appear as focal areas of a color deficit in the color variance image (as shown

later in an exemplary case). The RF data analysis was performed on these hypoechoic areas with color deficits to calculate the shear wave speed at different vibration frequencies.

In each participant, the most symptomatic site (ie, most painful active site) was then studied to measure the shear wave speed (Figure 1A). The myofascial trigger points were centered in the field of view before acquiring shear wave data. In the event that the participant had no myofascial trigger points, either the left or right upper trapezius was analyzed to acquire values for normal asymptomatic muscle tissue. Only active and normal participant data are presented in this study; latent trigger points were excluded because of the very small sample size.

Statistical Analysis

Shear wave speed values were aggregated for active myofascial trigger points, surrounding muscle tissue of active sites, and asymptomatic upper trapezius in normal participants. Statistical analysis was done using PASW version 18 (SPSS Inc, Chicago, IL) performing a 2-way analysis of variance (ANOVA) with a Tukey post hoc test. A 2-way ANOVA was performed to examine the effect of the tissue type (active trigger point versus normal and surrounding muscle tissue versus normal) and excitation frequency on the shear wave speed. A paired Student *t* test was performed between active trigger points and surrounding muscle tissue because these data were collected from the same muscle, and independence between these variable cannot be guaranteed. For all tests, the dependent variable shear wave speed was normally distributed for the groups formed by the combination of the tissue type or neck pain type and vibration frequency as assessed by the Shapiro-Wilk test. There was homogeneity of variance between groups, as assessed by the Levene test for equality of error variances. Statistical significance was determined as $P < .05$.

Results

In Vitro Validation Using the Experimental Phantom

On B-mode imaging, type I and IV spheres were isoechoic with the background material (Figure 2, C and E). Phase plots of the background alone (Figure 2B) show phase lag congruity throughout with a uniform spatial phase gradient. Substantial shear wave disruption was observed in the presence of both less stiff material (Figure 2D) and stiffer material (Figure 2F). The background material shear modulus was consistently measured whether alone or in the presence of type I or IV spheres (mean $\pm$ SD, 7.8 ± 1.8 , 6.7 ± 3.1 , and 9.2 ± 2.3 kPa, respectively). The shear moduli for type I and IV spheres were 2.5 ± 1.6 and 19.9 ± 2.0 kPa. Figure 4, top panel, shows the Voigt model fit to the shear wave speeds that was used to determine the shear moduli shown in Figure 4, bottom panel. These values were comparable to the manufacturer's specifications for the shear modulus ($G = 8.2$ kPa for background, 2.9 kPa for type I, and 24.9 kPa for type IV materials). Our method produced reproducible results with a low coefficient of variation for the background material and type I and IV spherical lesions (0.2, 0.69, and 0.27).

In Vivo Validation in Biceps Brachii

B-mode images of the long head of the biceps brachii in the healthy volunteers revealed homogeneous fiber alignment in the area of interest (Figure 3A). Phase images (Figure 3B) indicated fairly uniform phase lag with a uniform spatial gradient along the tissue for all vibration frequencies. The average phase lag over the imaged area was calculated as the slope of the linear fit of the phase angle and distance along the bicep. Figure 5 depicts data from an aggregate of 10 measurements collected from 5 participants and the resulting model fit used to estimate shear modulus and viscosity values. All shear wave speeds measured via US elastography were within the range of reported literature values (Figure 5, horizontal

lines). There was agreement between our data and those published in a previous study.⁸ However, in that study,⁸ the biceps muscle was assumed to be a nondispersive medium, as reported by Deffieux et al.¹⁰ Our preliminary measurements show dispersion of the shear wave with the frequency in the upper trapezius muscle. However, because of the small number of participants involved in this study and because of the difficulties involved in achieving a perfect alignment between muscle fibers and the wave direction, further studies are needed to confirm whether the biceps muscle shows dispersion. Measured shear values for the biceps brachii were highly repeatable, with differences between run 1 and run 2 not being greater than 5 kPa.

Preliminary Results in the Normal Upper Trapezius Versus Trapezius With Active Myofascial Trigger Points

B-mode images of the normal upper trapezius show a uniform fiber orientation throughout the tissue (Figure 3C). The phase of the shear wave propagating in the lateral direction showed a relatively uniform spatial gradient throughout the muscle (Figure 3D). Participants with active trigger points showed spherical or bandlike hypoechoic (darker) regions along with an increase in fiber alignment heterogeneity (Figure 3E), as reported previously by our group.^{24,25} Phase images showed a heterogeneous pattern of spatial phase gradients compared to normal tissue (Figure 3F). The phase plots between normal and active participants showed that not only is there a difference in the overall phase lag of the muscle at a given frequency, but there is also localized shear wave disruption at the site of the myofascial trigger point. The observed shear wave disruption is similar to what was observed in the phantom experiments with stiffer inclusions (Figure 2). Figure 6 shows an exemplary case of a patient with a palpable active myofascial trigger point. The B-mode image shows hypoechoic regions corresponding to the regions of the palpated trigger point, whereas color variance imaging during a 50-Hz vibration shows regions of a color deficit, consistent with our previous findings.^{24,25} High-frame rate imaging at 532 frames per second was then performed at the location of the myofascial trigger point, and the raw RF data were processed to yield shear speed images. An image of the distribution of the shear speed at 179 Hz shows a local increase in the shear wave speed corresponding to the active myofascial trigger point.

From the spatial phase gradients, shear wave velocities were calculated for normal tissue, active myofascial trigger points, and their surrounding muscle tissue. Both active trigger points and the surrounding muscle tissue were significantly different in terms of shear wave speeds (at frequencies >100 Hz) than normal tissue in the upper trapezius ($P < .05$; Figure 7). At low frequencies (350 Hz), there were no significant differences in the shear wave speed. Although there were some differences in shear moduli using Voigt model fits between active trigger points and surrounding muscle, there was large variability in the dispersion, making the estimated model parameters less reliable. When the shear speeds were aggregated for all cases, there were no significant differences in the shear wave speed between active trigger points and the surrounding muscle tissue at different frequencies. The large spatial variability in the shear wave speed in symptomatic patients indicates that the neighborhood of active myofascial trigger points is heterogeneous. In contrast, shear wave speeds of normal tissues were well clustered along a Voigt model fit with low variability, with an average shear modulus of 3.38 kPa.

Discussion

This study presents a new US method, suitable for clinical settings, to quantitatively measure mechanical shear properties of superficial soft tissues in the body via US elastography. This method provides quantitative estimates that can be used clinically to identify and determine the location of active myofascial trigger points and changes in their

tissue properties over time, including their response to treatment, and can overcome the subjectivity of the physical examination. We have shown that with the use of a superficial handheld vibrator, it is possible to estimate the shear modulus and shear viscosity of muscle tissue 0 to 4 cm from the skin's surface conveniently in an office-based setting.

Several approaches to US elastography have been proposed previously in the literature, and many excellent reviews of the main methods have been published (eg, Greenleaf et al²⁸). These methods fall mainly into two categories: static elastography or strain imaging²⁹ and dynamic elastography. Static methods involve the US measurement of the strain induced by the application of compressive stress on tissue. Strain imaging can provide qualitative images of stiffness distribution, but it cannot directly provide quantitative estimates because the stress distribution is unknown. Dynamic methods typically involve some perturbation of the tissue, either using an external vibration (such as sonoelastography³⁰) or the radiation force of US (such as acoustic radiation force imaging,⁸ vibroacoustography,³¹ and supersonic shear wave imaging³²). The shear waves induced by these perturbations can then be imaged with US, and the amplitude and speed of the shear wave can be related to the mechanical properties. Shear wave approaches have become popular recently because of the possibility of quantitative characterization. The advantage of our technique compared to other shear wave elastographic methods previously proposed in the literature is the portability and accessibility of the equipment. This factor is particularly important for musculoskeletal examinations, which are often office based. Methods based on acoustic radiation force or supersonic shear wave elastography^{33,34} require specialized US systems that are not readily available. In contrast, our proposed method uses off-the-shelf components (a variable-voltage power supply, high-revolutions-per-minute direct current motor, and commercially available clinical US system with a research interface that is capable of accessing the raw RF data). The objective of this study was to evaluate whether the proposed method can produce reliable quantitative estimates of mechanical properties of skeletal muscle that are of clinical relevance.

To validate our shear wave elastographic system, we performed both in vitro tests using a calibrated experimental phantom containing spherical lesions of varying stiffness embedded in a background material and in vivo tests in the biceps brachii to replicate literature values determined using previously proposed methods.

The results of the phantom experiments showed that our proposed method can quantitatively estimate the shear modulus of 3 types of materials of different stiffness (background and 2 types of inclusions) consistently with good reproducibility. The estimated shear moduli match favorably with the manufacturer's specified values, with absolute errors of 0.4, 0.4, and 4.9 kPa for mean shear moduli for background, type I, and type IV materials, respectively, which correspond to 5%, 13%, and 19% error rates. These small differences in estimated shear moduli could have been due to interference caused by shear wave reflection from the walls encasing the phantom and also the applicability of the Voigt model, which assumes a homogeneous viscoelastic material. Attempts were made to minimize errors caused by shear wave reflection from the phantom enclosure by limiting the data capture to within the first 2 seconds of onset of vibration and then only using a 225-millisecond window, which shows shear wave progression without any reflective interference. However, the shear wave reflection interference was not observed in vivo in the biceps brachii or upper trapezius experiments. Another contributor to the estimation error could have been the boundary and loading conditions from interactions between the stiffer and softer inclusions with the background medium. The shear moduli measurements by the manufacturer were performed for each individual material directly before assembly. The interactions of the surrounding material when performing US elastography could explain some of the differences in the estimated values.

For the biceps experiments, all of our estimated values were within the range of previously published literature values of shear speed for the biceps brachii (1.8–3.9 m/s).^{5,10} These results confirmed that our office-based technique was capable of replicating elastographic measurements reported using magnetic resonance elastography¹⁵ or supersonic pulse waves,^{5,10} and we could measure shear properties in a relatively homogeneous muscle with uniform fiber alignment.

To show the clinical applicability of our method *in vivo*, we compared the shear wave velocities of the upper trapezius in asymptomatic healthy volunteers without trigger points to those of symptomatic patients who had neck pain and tissue with palpable active myofascial trigger points. Myofascial trigger point regions were identified for shear wave analysis using two methods that we have previously described.²⁶ Palpable myofascial trigger points appear as focal hypoechoic regions on B-mode images and as regions of a focal color deficit on color variance images when the muscle is vibrated with an external vibrator. Our results show that in muscles with an active myofascial trigger point (ie, spontaneous cervical pain), the shear wave traveled significantly faster than in palpably normal muscles in asymptomatic participants when the excitation frequency was higher than 100 Hz, whereas it did not for low-frequency excitation (≈ 50 Hz). The variance of the symptomatic cases was also higher than that of the normal controls. These findings suggest that the soft tissue milieu of active myofascial trigger points is substantially stiffer and more heterogeneous than normal tissue.

The presented technique provides an objective quantitative measure of muscle properties that can be used in the future for clinical assessment and for measuring treatment outcomes in conjunction with digital palpation by a skilled clinician. Digital palpation does not do the following: (1) provide an objective, reliable, and sensitive method of diagnosis and measurement of treatment efficacy; (2) provide quantitative comparisons of the tissue properties before and after treatment; (3) objectively differentiate among active myofascial trigger points, latent trigger points, and palpably normal tissue; (4) objectively discriminate between superficial and deep trigger points; and (5) permit an objective study of the natural history of the trigger points. We have previously shown that US imaging can identify myofascial trigger points and differentiate between small and large as well as superficial and deep trigger points.^{25,26} In this study, we have further shown that the viscoelastic properties of the muscle in symptomatic patients are quantitatively different from those of palpably normal muscle, which could provide a method for evaluating treatment efficacy in the future. Ultrasound elastography is well suited for long-term monitoring of painful soft tissue conditions such as myofascial pain syndrome. Longitudinal monitoring of the shear modulus of both myofascial trigger points and the surrounding muscle tissue over time before and after treatment may provide objective natural history data that will provide a better understanding of the changes that occur in the mechanical environment of the upper trapezius after treatment or intervention for myofascial trigger points, such as dry needle therapy.³⁵ Our research group is currently investigating these changes in an ongoing study.

To further investigate the clinical applicability, we are investigating the differences in shear wave speeds between patients who have acute onset of neck pain (<3 months) versus those with chronic neck pain (>3 months). In a preliminary analysis of the small number of participants recruited for this study, we observed significantly higher shear wave speeds in patients with chronic neck pain for frequencies higher than 130 Hz ($P < .04$), whereas there was no significant difference at lower frequencies. This finding suggests that the soft tissue milieu in acute neck pain is less viscous compared to that in chronic neck pain. This finding could explain some of the variability in the data observed in Figure 7. Of course, these findings are preliminary and need to be investigated further.

At present, our proposed method has been tested for superficial applications (<4 cm deep) to achieve high imaging frame rates for tracking the propagating shear wave. However, on the basis of measurements taken from the phantom, biceps brachii, and normal upper trapezius, shear property measurements do not appear to be dependent on depth within the 4-cm measuring range we propose. The proposed method would be suitable for most musculoskeletal applications; however, the applicability of this method for deeper tissue needs to be investigated in the future.

Although this study focused on the most prominent and symptomatic myofascial trigger point present in a given patient, our technique should be able to identify small, deeply located trigger points that are stiffer than the surrounding muscle tissue. We have previously shown that latent trigger points are smaller (0.27 cm^2) compared to active trigger points ($0.35\text{--}0.40 \text{ cm}^2$),²⁶ which are well within the spatial resolution of our technique.

The technique is sensitive to the alignment of the US probe and to the position of the handheld vibrator to achieve proper coupling of the shear wave in the muscle. The shear wave can be missed if the vibrator and probe are not aligned and placed correctly. The alignment problem was addressed by providing interactive visualization of the phase image, allowing the user to adjust the position of the probe and vibrator to ensure the shear wave propagation. Data collected from in vivo experiments were affected by a large standard deviation (sometimes >50% of the mean value) compared to data obtained from the phantom. This large variance could have been due to the anisotropy of the muscle and the difficulty in aligning the US probe to be parallel to the muscle fibers. The Voigt model, typically used to interpret the shear modulus from the dispersion of shear wave speeds, is sensitive to variance; ie, a small variation in shear wave speed measurements can cause a large variation in the shear modulus and viscosity. In our analysis, we have therefore reported the raw shear wave velocities instead of the Voigt model for our in vivo data. Another limitation of our study was the inability to measure shear properties of skeletal muscle perpendicular to the fiber direction. The mechanical properties of muscle are known to be highly anisotropic.^{36,37} In this study, shear properties were only estimated parallel to the fiber direction. It was found that shear waves traveled much better along fibers in the parallel direction compared to the perpendicular direction. Shear waves from superficial vibrations were rapidly absorbed when traveling perpendicular to the fiber direction (data not shown).

This study presents a new US elastographic method that measures shear properties of superficial soft tissues. The method was validated through in vitro and in vivo studies and shows the feasibility of measuring shear properties of skeletal muscle tissue using standard US scanning equipment and techniques. A clinical application of this method is to provide an objective assessment of soft tissue and the identification of myofascial trigger points. This method may have important applications for diagnosis and treatment evaluation of myofascial trigger points and myofascial pain syndrome.

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Abbreviations

ANOVA analysis of variance

RF	radio-frequency
US	ultrasound

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Figure 1.

A, Vibrating upper trapezius with the office-based custom vibrator. **B,** Disassembled view of the plastic housing and direct current motor with an offset weight attached to its rotor axis to generate vibrations. The white screw knob fixes the motor in the plastic housing.

Figure 2.

B-mode (gray-scale) and phase plot images of shear wave vibrations at 160 Hz for the phantom background material (**A** and **B**), type I sphere with surrounding background material (**C** and **D**), and type IV sphere with background material (**E** and **F**). Numbered lines denote areas where phase lag was measured. Circled areas denote type I and IV sphere locations.

Figure 3. B-mode (gray-scale) and phase plot images of shear wave vibrations at 160 Hz for the biceps brachii (**A** and **B**), normal upper trapezius (**C** and **D**), and upper trapezius with an active myofascial trigger point (MTrP; **E** and **F**). Numbered lines denote areas where phase lag was measured. Circled areas denote the active trigger point.

Figure 4.

Top, Shear wave speeds (points) and model fits (lines) for phantom gel studies. Bottom, Average estimated shear modulus for the background (BG) material, type I sphere, type IV sphere, and surrounding background material for both spheres.

Figure 5.

Phase shear wave speeds for the biceps brachii. Blue points indicate data collected during this investigation, and the Voigt model fitting (blue line) shows dispersion of the shear wave (5.11 kPa and 6.9 Pa/s). Gray, orange, and red lines denote reported literature range values for the biceps shear speed when the shear wave propagates along the fibers. Green lines denote reported literature range values for the biceps shear speed when the shear wave propagates orthogonal to the fibers.

Figure 6.

A and B, Myofascial trigger points observed as hypoechoic regions on a B-mode image (**A**) and as regions of a color deficit on a color variance image (**B**) in a patient with acute neck pain and a palpable active myofascial trigger point. The red arrows highlight two hypo - echoic areas in the B-mode image, which are vibrating with a lower amplitude than the surrounding tissue, as shown in the color variance image, leading to the color deficit. **C and D,** B-mode image (**C**; low resolution, with reduced line density, 532 frames per second) and shear wave speed at 179 Hz overlaid on the B-mode image (**D**). There is an increase in the local shear wave speed in the center of the image corresponding to the palpable active trigger point (white arrow). The B-mode image shows the corresponding hypoechoic region (white arrow). The upper trapezius muscle is the region shown by the double-sided arrow. The imaging depth is 25 mm in all images.

Figure 7.

Shear wave speed for the upper trapezius. Red points indicate normal tissue; blue points, active myofascial trigger point (MTrP); and green points, surrounding muscle tissue (SMT) around the trigger point. Lines are Voigt model fits showing dispersion of the shear wave within the muscle tissue. The green model fit partially overlaps the blue. Shear modulus and viscosity values used for fitting the data were as follows: active trigger point, 3.5 kPa and 10.8 Pa/s; normal, 3.38 kPa and 6.09 Pa/s; and surrounding muscle tissue, 2.14 kPa and 10.1 Pa/s.