Assessment of individual muscle hardness and stiffness using ultrasound elastography

Takayuki Inami* and Yasuo Kawakami

Faculty of Sport Sciences, Waseda University, 2-579-15 Mikajima, Tokorozawa, Saitama 359-1192, Japan

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Abstract Assessing muscle mechanical properties such as hardness and stiffness has important clinical implications. The use of ultrasound elastography to assess individual muscle hardness or stiffness has been increasing in recent years. Several different ultrasound elastography methods are currently in use, including strain elastography and shear wave elastography, which are capable of capturing the distribution of hardness and stiffness within individual muscles. In the present review, we outline some of the current basic and clinical applications of strain elastography and shear wave elastography, and illustrate how such ultrasound elastography technologies may further advance understanding of individual muscle mechanical properties as well as muscle functions.

Keywords: strain elastography, shear wave elastography, muscle force, static stretching

Introduction

Muscle hardness and stiffness is affected by conditions such as spasms, edema, contracture, swelling, and compartment syndrome. The dysfunctional muscle tissue often demonstrates abnormal mechanical properties1-5), which highlights the clinical importance of obtaining accurate information about individual muscle mechanical properties in relation to physiological and pathophysiological conditions. Nevertheless, muscle mechanical properties such as hardness and stiffness have traditionally been assessed through indirect methods such as palpation, manual muscle testing, and measurement of muscle strength and joint range of motion. While such assessments have provided valuable information and played a fundamental role in the physical examination of patients and research subjects, their objectivity is limited. Moreover, such assessment approaches are unable to differentiate the mechanical properties of individual muscles from those of the associated tendons, neurovascular structures, joint capsule, subcutaneous fat, or skin.

The use of ultrasound elastography to assess muscle hardness and stiffness has been increasing in recent years. Several different ultrasound elastography methods are currently in use, including strain elastography (SE) and shear wave elastography (SWE). These ultrasound-based techniques are capable of capturing the distribution of hardness and stiffness within individual muscles, and thus show promise for direct measurement of the mechanical properties of individual muscles. In the context of ultrasound elastography measurements, muscle mechanical properties are typically reported as “muscle hardness” or “muscle stiffness”. However, these terms are often used ambiguously, as muscle hardness is associated with muscle stiffness6). Therefore, before we proceed with our descriptions, the definition of these terms should be clarified. Muscle hardness has been defined as the resistance offered by the muscle against perpendicular pressure4,7,8), while muscle stiffness has been defined as the ratio of change in force to change in length along the muscle’s longitudinal axis9,10). Therefore, muscle hardness is a mechanical property that represents transverse muscle stiffness, and is distinguished from the general term muscle stiffness, which is, by definition, measured along the longitudinal axis of a muscle. In this context, it is important to note that SE measures strain that represents muscle hardness in a direction perpendicular to muscle shortening, while SWE generally measures muscle stiffness (shear modules) along the muscle shortening direction11). Moreover, it is also important to understand how changes in muscle hardness and stiffness are reflected in the values obtained from ultrasound elastography. In this short review, we provide a brief overview of these two most common ultrasound elastography approaches (SE and SWE) and outline some of their current basic and clinical applications.

Strain elastography

In SE (sometimes referred to as compression or real-time tissue elastography), the operator manually compresses the ultrasound transducer (the rhythmical compression-relaxation cycles) against the patient’s body surface. Subsequently, a qualitative elastogram (elasto-
graphic image) is constructed based on the principle that, compared to harder tissue, softer tissue has more deformation and therefore experiences larger strain. To enable a more objective assessment, this qualitative SE data is typically converted to a semi-quantitative form as the strain ratio between the strain of a region of interest (ROI) in the target muscle and that of a reference material (e.g., an acoustic coupler) on the same elastogram. The strain ratio is calculated by the following formula:

$$\frac{E_{\text{muscle}}}{E_{\text{reference}}} = \frac{\frac{\sigma_{\text{muscle}}}{\epsilon_{\text{muscle}}}}{\frac{\sigma_{\text{reference}}}{\epsilon_{\text{reference}}}}$$  \[\text{Eq. 1}\]

where \( E \) represents Young’s modulus, \( \sigma \) represents stress, and \( \epsilon \) represents strain. \( E \) is given as stress over strain (\( \sigma / \epsilon \)), according to Hooke’s law.

The calculation of muscle hardness using strain measurement is based on the assumption that the stress field is homogeneous. Depending on the SE device, it is possible to calibrate the strain by pressing the probe against the muscle and against a reference material repeatedly during the measurement, generating a strain graph or strain bar. It has been indicated that strain ratio values are reliable only if the reference strain is constant during the calibration; therefore, the operator should always perform calibration to ensure a reliable measurement.

SE has been applied in the field of sports science for measuring the change in muscle hardness in the biceps brachii, rectus femoris, and gastrocnemius muscles. One advantage of SE is that it can secure a relatively large ROI. It has been suggested that the hardness of the relaxed muscle depends on muscle size, and thus it is important to be able to set a larger ROI in order to accurately assess larger muscles. To the best of our knowledge, ROIs of 600 mm$^2$ can be reliably assessed by SE, whereas such a ROI size cannot currently be assessed by SWE, as we describe in detail in the next section. In addition, it has been recently reported that strain ratio is not influenced by depth. Thus, SE is more suitable for assessing hardness of the entire muscle, or the hardness of muscles located at a greater depth. However, since the reliability of SE measurements is affected by the characteristics of the rhythmic compression-relaxation cycles, SE depends on the skills of the operator.

Shear wave elastography

SWE is an ultrasound-based technique that uses shear waves to measure tissue stiffness in a quantitative manner. SWE evaluates muscle stiffness in absolute values by using an ultra-high-speed algorithm that computes Young’s modulus of elasticity based on the speed of shear waves generated from the transducer. Thus, within SWE, the shear wave propagation velocity is directly related to the shear modulus of the tissue, that is, the stiffer the tissue, the faster the shear wave propagation:

$$\mu = \rho \cdot V_s^2$$  \[\text{Eq. 2}\]

where \( \mu \) is the shear modulus of the tissue, \( \rho \) is the density of the muscle (1055 kg/m$^3$), and \( V_s \) is the shear wave velocity.

To our knowledge, the supersonic shear wave imaging (SSI) apparatus (Imagine, Aix en Provence, France) represents the current state-of-the-art device in ultrasound elastography, because it provides quantitative data regarding individual muscle stiffness. Unlike the semi-quantitative results provided by SE, SWE provides quantitative data in absolute values, but the ROI in SSI is much smaller. However, as new techniques are being developed, this limitation may soon be overcome; indeed, recent studies reported the successful use of larger ROIs (up to 400 mm$^2$). It has been suggested that the ROI should be as large as possible for accurate measurement of muscle stiffness. On the other hand, SWE is considerably less operator-dependent than SE, as tissue deformation relies on highly controlled ultrasound push beams rather than on tissue compression by the sonographer. Moreover, according to the work of Kot et al., compression of the tissue under the ultrasound transducer can increase tissue stiffness, an effect related to the nonlinearity of the tissue rather than to measurement bias. In this context, SWE provides an advantage over SE, since SWE requires only minimal compression, amounting to merely maintaining transducer contact with the skin surface.

Both SWE and SE assume that the underlying tissue is isotropic, elastic, and locally homogenous, as is the case for breast, liver, or thyroid tissue. Muscle tissue, however, is highly anisotropic, and the mechanical properties along the muscle fibers differ from those across the fibers. For this reason, the transducer must be oriented longitudinally to the muscle fibers in order to achieve accurate and reliable SWE measurements. Despite anisotropy, a swine model of the brachialis indicated that SWE-derived shear modulus values were highly correlated with those of the Young’s modulus measured by a conventional stress–strain test. However, image stability and values based on SWE images varied with the imaging plane (i.e., transverse and longitudinal), and thus the imaging plane should be taken into consideration when measuring skeletal muscle tissue elasticity by SWE. Additionally, it has been suggested that shear modulus measured by SWE may be considerably affected by the pennation angle, although more recent reports indicate that this effect is negligible (<1.3%). Thus, compared to SE, SWE has been found to be more useful in basic investigations, and its use is expanding.

The relationship between individual muscle stiffness (shear modulus) and muscle contraction intensities

SWE has been applied to assess the relationship between muscle contraction intensity and muscle stiffness. Nordez and Hug showed that the SWE-derived elastic modulus of the biceps brachii muscle increased linearly
with the contraction intensity that ranged from 0 to 40% of maximal voluntary contraction (MVC). Similarly, Yoshitake et al. 30 reported that the biceps brachii muscle became 4 times stiffer with an increase in torque from 15% to 60%, and the slope of the stiffness increase was associated with an increase in electromyographic activity. Ateş et al. 22 have recently shown that the shear modulus of the abductor digiti minimi muscle increases linearly with contraction intensity ranging from 0% to 100% of MVC. In other words, these results suggest that the shear modulus of individual muscles can be used to reliably estimate the force produced by an individual muscle 21,31,32. Evidence from studies that did not use ultrasound elastography (e.g., using push-in meter) suggests that the increase in muscle stiffness for contractions with up to moderate intensity is mainly due to the number of cross-bridges 33,34. However, push-in meters have indicated that the increase in muscle hardness is blunted at contractions with higher intensity 35. As muscle fibers fuse at contractions of higher intensity, muscle force is increased by an increase in firing rate 36. Morisada et al. 35 concluded, since muscle hardness during tetanic nerve stimulation was closely correlated with the M-wave amplitude, that the contractile activity (interactions of cytoskeletal filaments accompanied by neuromuscular transmission) and muscle membrane excitability represent mechanisms likely related to the development of muscle hardness. Therefore, currently, the conclusions based on SWE-derived data contrast the long-held hypothesis regarding motor-unit recruitment strategies. Unfortunately, there has been no SE-based investigation of the relationship between muscle hardness and the intensity of muscle contraction.

**Clinical applications**

Many rehabilitation strategies are aimed at changing the mechanical properties of the muscle. However, these changes cannot be measured directly and reliably in the clinical setting. Ultrasound elastography techniques provide noninvasive, quantitative (or semi-quantitative) and reliable measurements; most importantly, individual muscle hardness and stiffness can be assessed independently. Such characteristics represent the great advantages of ultrasound elastography techniques for clinical assessment and rehabilitation purposes. As common therapeutic strategies including stretching, massage, and taping aim to alter muscle hardness and stiffness, the individual assessment of individual muscle hardness and stiffness may also be helpful to quantify the efficacy of new therapies.

**Stretching.** Several previous studies have used the SWE-based shear modulus (i.e., muscle stiffness) to investigate the stretching effect on hamstrings and triceps surae muscles. Miyamoto et al. 37 have recently examined whether the effects of hamstring stretching on the passive stiffness of the long head of the biceps femoris, semimembranosus, and semitendinosus vary with the type of stretching maneuver (i.e., passive knee extension and hip flexion). They found that, for a given knee joint angle, passive knee extension significantly reduced the shear modulus in the long head of all three muscles investigated. Interestingly, passive hip flexion had a significant stretching effect only in the semitendinosus and semimembranosus. The authors therefore concluded that, in terms of reducing the muscle stiffness of the long head of the biceps femoris, stretching of the hamstring should be performed by passive knee extension rather than by hip flexion. Nakamura et al. reported that the stiffness of the gastrocnemius muscle decreased by 19% and 36% after short static stretching (SS) sessions on a torque machine, consisting of 4 sets of 30 seconds each (total 2 min) and 5 sets of 60 seconds each (total 5 min), respectively 38,39. These results suggest that the duration of SS influences the degree of decrease in muscle stiffness. Taniguchi et al. 40 showed that the stiffness of the gastrocnemius muscle decreased by 14% after SS consisting of five 1-minute repeats of standing wall stretching in dorsiflexion. They also reported that the shear modulus recovered within 20 minutes after stretching. Hirata et al. 41 examined the muscle stiffness response of the medial gastrocnemius, lateral gastrocnemius, and soleus during passive dorsiflexion before and after SS; they found that the stretching induced significant reductions in shear modulus of the medial gastrocnemius, but not in that of the lateral gastrocnemius or soleus. This result indicates that passive muscle stiffness differs among the triceps surae. Akagi et al. 42 reported that the stiffness of the gastrocnemius muscle decreased by 13% after SS consisting of three 2-minute repeats (6 minutes in total) of standing wall stretching in dorsiflexion. These authors later demonstrated 43 that muscle stiffness of the gastrocnemius decreased by 7-10% following a 5-week SS program consisting of three 2-minute repeats. All the reports described above on stretching effects are based on SWE investigations. To our knowledge, our study 44 is the only investigation that employed SE when assessing muscle hardness. Inami et al. 41 found a similar decrease (12-18%) after SS consisting of three 2-minute repeats, to those in SWE-based studies 38,42. We feel that it is important to systematically compare between the results of SE and those of SWE performed with the same experimental setting to better understand the relationship between SE- and SWE-derived data.

**Massage and taping.** Massage is another technique that may be used to reduce muscle hardness and stiffness. In their SWE-based assessment, Eriksson Crommert et al. 45 provided the first direct evidence that a 7-minute session of massage effectively decreases muscle stiffness of the gastrocnemius, though this effect did not persist after a short period of rest (3 minutes). However, it should be noted that the study of Eriksson Crommert et al. involved asymptomatic participants. In addition to the fact that the
effectiveness of massage is influenced by the technique applied and skill of therapist, it remains unclear whether the clinical effectiveness of massage techniques can be accurately assessed by SWE.

Hug et al. demonstrated that deloading tape applied to the skin directly over the rectus femoris muscle reduced tension in the underlying muscle region during moderate and high muscle stretch and contraction. This result provided a biomechanical explanation for the effect of deloading tape observed in clinical practice, and such assessments may be used for monitoring athletic performance and injury prevention.

Conclusions

Ultrasound elastography provides the opportunity to further our understanding of the interaction between muscle structure and function by allowing measurement of the mechanical properties of individual muscles.

Conflict of Interests

The authors declare that there is no conflict of interests regarding the publication of this article.

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