Current status of musculoskeletal application of shear wave elastography

ULTRASONOGRAPHY

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REVIEW ARTICLE

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Ultrasonography (US) is a very powerful diagnostic modality for the musculoskeletal system due to the ability to perform real-time dynamic high-resolution examinations with the Doppler technique. In addition to acquiring morphologic data, we can now obtain biomechanical information by quantifying the elasticity of the musculoskeletal structures with US elastography. The earlier diagnosis of degeneration and the ability to perform follow-up evaluations of healing and the effects of treatment are possible. US elastography enables a transition from US-based inspection to US-based palpation in order to diagnose the characteristics of tissue. Shear wave elastography is considered the most suitable type of US elastography for the musculoskeletal system. It is widely used for tendons, ligaments, and muscles. It is important to understand practice guidelines in order to enhance reproducibility. Incorporating viscoelasticity and overcoming inconsistencies among manufacturers are future tasks for improving the capabilities of US elastography.

Keywords: Ultrasonography; Elasticity imaging techniques; Tendinopathy; Muscles; Elasticity

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Introduction

As a result of recent advances in ultrasonographic technology, musculoskeletal ultrasonography (US) has become increasingly common in the past decade [1]. Due to its widespread accessibility and relatively low cost, as well as the inherent possibility of real-time dynamic examinations using US, US is an irreplaceable modality in the musculoskeletal field [2,3]. The pixel size of the most recent high-frequency probe is as small as one-third of that of 1.5-T magnetic resonance imaging (MRI), so small superficial structures such as tendons, ligaments, and subcutaneous tissues can be better evaluated with US at a higher level of spatial resolution [3,4]. Doppler US can detect hyperemia in soft tissue structures [2,3].

Since US has the inherent limitation of not being able to show the biomechanical properties of tissues, it has been difficult to assess the relationship between structural disorganization and clinical pain [5]. Now, with shear wave elastography, in addition to obtaining morphological information, we can quantify the absolute elasticity value of soft tissue structures and obtain useful quantitative information about the mechanical properties related to degeneration, injury, and healing [5]. This technology is the US-based counterpart to the palpation usually done manually by clinicians to diagnose and characterize tissue [6,7].

Shear wave elastography involves the following processes: generation of shear waves in tissue by an acoustic radiation force, detection of the propagation of the induced shear waves, and processing of the shear waves to create elastograms, which are quantitative maps of tissue elasticity [6,7].

US elastography has been established as an excellent diagnostic method for liver fibrosis, breast cancer, and thyroid cancer [8-14]. In the musculoskeletal field, much research has been conducted on US elastography since the early 1990s [15], and recently it has been applied to clinical practice [16].

The purpose of this review article is to introduce this technology to musculoskeletal radiologists and to address the current status of US elastography for the musculoskeletal system with an emphasis on shear wave elastography.

Basic Concepts

Effective Differentiation of Tissue Using the Elastic Modulus

The tissue characteristic evaluated by manual palpation is known as the elastic modulus in engineering terms [7]. Computed tomography utilizes the attenuation coefficient of tissue to visualize the spectrum of the contrast mechanism, and MRI uses the T1 relaxation time to show tissue contrast. Conventional US uses the bulk modulus. Shear wave elastography uses the shear modulus, and the shear modulus shows the greatest variation, with over five orders of magnitude among various physiological states of normal and pathologic tissue [7]. This means that the use of the elastic modulus can allow sensitive visualization of the differences in biomechanical properties among tissues (Fig. 1) [10]. This is the reason why musculoskeletal radiologists can use US elastography to diagnose the compromised

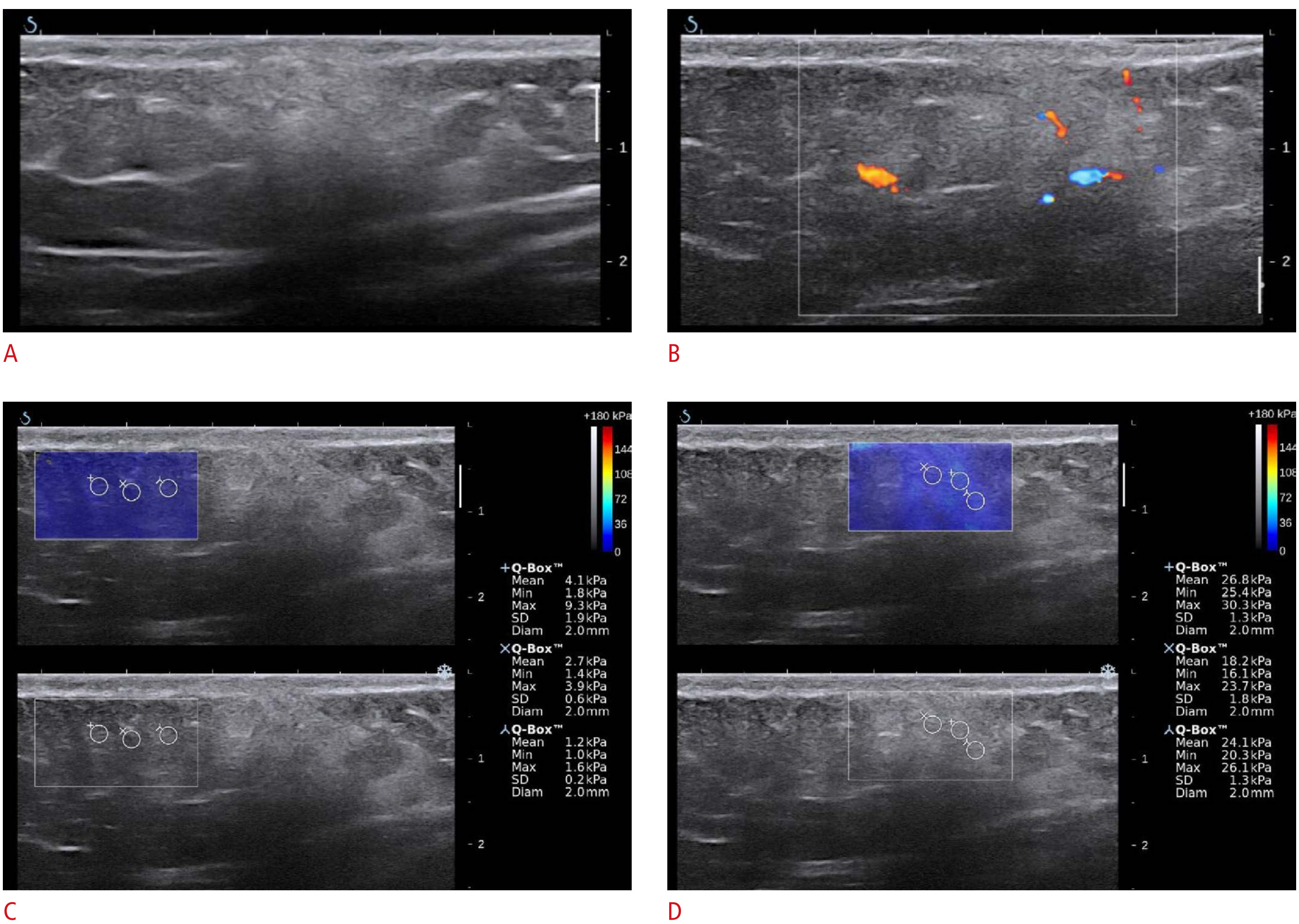
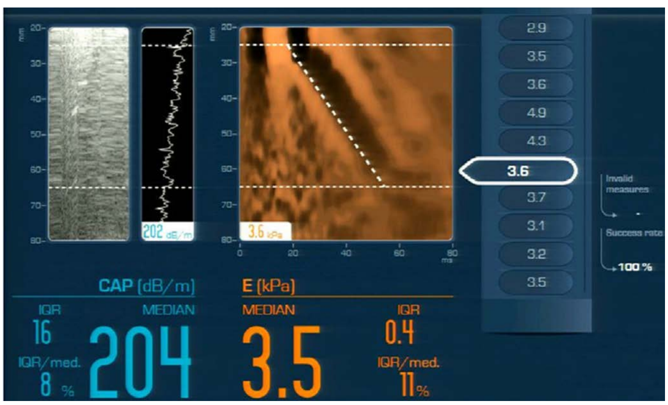


Fig. 1. Shear wave elastography of a palpable superficial mass lesion in the right lower quadrant of the abdomen in an 18-year-old man.
A. In the grayscale image, an ill-defined hyperechoic soft tissue lesion with posterior acoustic shadowing was noted. B. The fatty lesion shows increased vascularity in a color Doppler image. C. In shear wave elastography, the adjacent normal subcutaneous fat tissue shows elasticity measured at 1.2-4.1 kPa. D. In the shear wave elastography, the lesion shows much higher elasticity, measured at 18.2-26.8 kPa.

tissues seen in degeneration or fibrosis [5].

Shear Wave Elastography as the Most Suitable US Elastography Method

Among the various commercially available US elastography devices, transient elastography does not provide a B-mode image, so the accurate anatomic targeting of elasticity measurement is impossible [10]. Acoustic radiation force impulse imaging produces a grayscale image; however, the region of interest (ROI) is small, and fixed at a depth of 4 cm [17]. Real-time strain elastography is the most commonly used method and can obtain the tissue elastogram and B-mode image simultaneously; however, it is operator-dependent,



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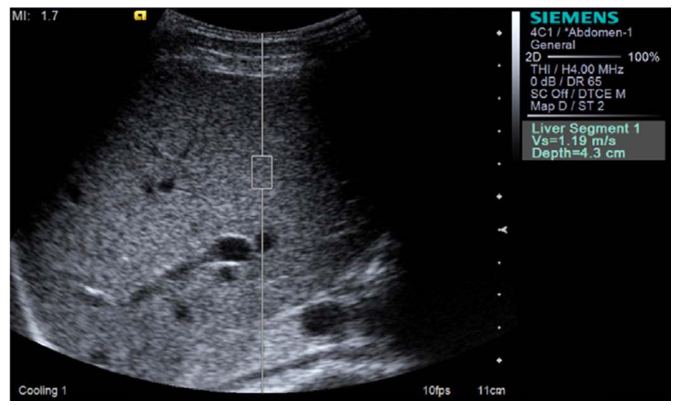

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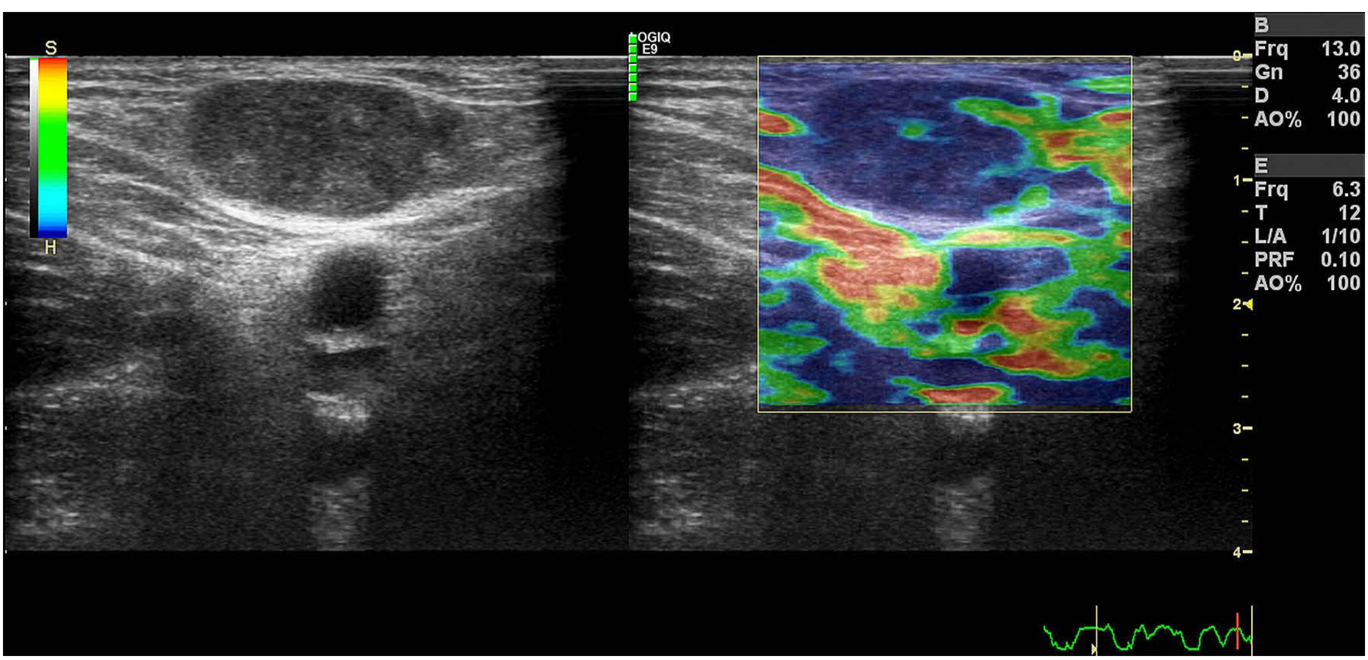

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Fig. 2. **Various techniques of ultrasound elastography.**
A. In a transient elastography of a normal liver, the displacement M-mode image located in the center of the monitor shows axial displacement as a function of depth (y-axis) and time (x-axis). Reprinted from Jeong et al. Ultrasonography 2014;33:149-160 [10], according to the Creative Commons licence Korean Society of Ultrasound. B. In an acoustic radiation force impulse imaging of a normal liver, the cylindrical region of interest in the middle of the ultrasonogram was used as the sample. Instead of the Young modulus, the propagating velocity of the shear wave is displayed. Reprinted from Jeong et al. Ultrasonography 2014;33:149-160 [10], according to the Creative Commons licence Korean Society of Ultrasound. C. Strain elastography enables real-time grayscale ultrasonography and a corresponding color map. Reprinted from Kim et al. Ultrasonography 2016;35:104-109 [19], according to the Creative Commons licence Korean Society of Ultrasound.

and cannot be used to calculate the absolute elastic modulus [18–20]. In addition, to obtain the relative strain ratio of tissue, real-time strain elastography needs a reference ROI, which is usually standard subcutaneous fat tissue with constant elasticity; however, in the musculoskeletal system, this is often difficult or impossible to accomplish for anatomic reasons (Fig. 2) [17,20]. Shear wave elastography is an operator-independent, relatively reproducible, and quantitative method useful for the evaluation of tendon and muscle [21], in spite of the size, shape, and depth limitations of the currently available ROI [17].

Assumptions of Shear Wave Elastography: Tissue Is Elastic, Homogenous, and Isotropic

Generally, soft tissues are viscoelastic, inhomogeneous, and anisotropic [22]. Viscoelastic tissues have both elastic solid properties and viscous fluid properties (Figs. 3, 4) [6,23]. For elasticity metrics, if the viscous forces are ignored, assuming linear, elastic solid tissues, a first-order approximation is possible [6,22]. All commercially available US elastography systems are based on the prerequisite assumption that the material is elastic, incompressible, homogenous, and isotropic [22,24]. In fact, the soft tissue elasticity in the human body is nonlinear and dependent on the tissue density, strain magnitude, and/or applied excitation frequency [6].

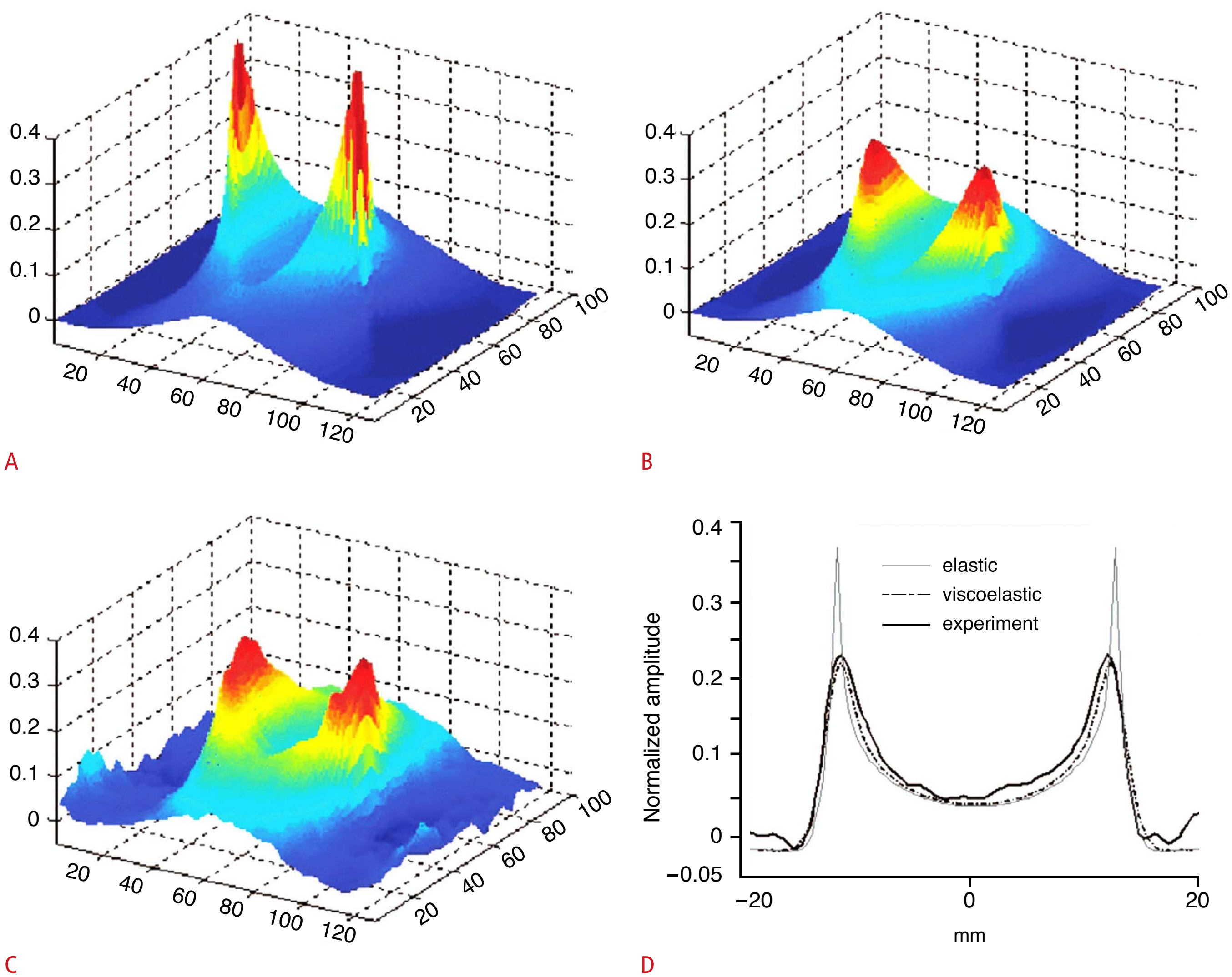

Fig. 3. **Purely elastic (A), viscoelastic (B), and experimental (C) 3-dimensional plots of the spatial shear wave pattern in the (x, z) plane at a given sampling time.**
Plot (D) represents the variation of those three fields along the x-axis (at z=0). Reprinted from Bercoff et al. IEEE Trans Ultrason Ferroelectr Freq Control 2004;51:1523-1536 [22], with permission from the IEEE Xplore Digital Library.

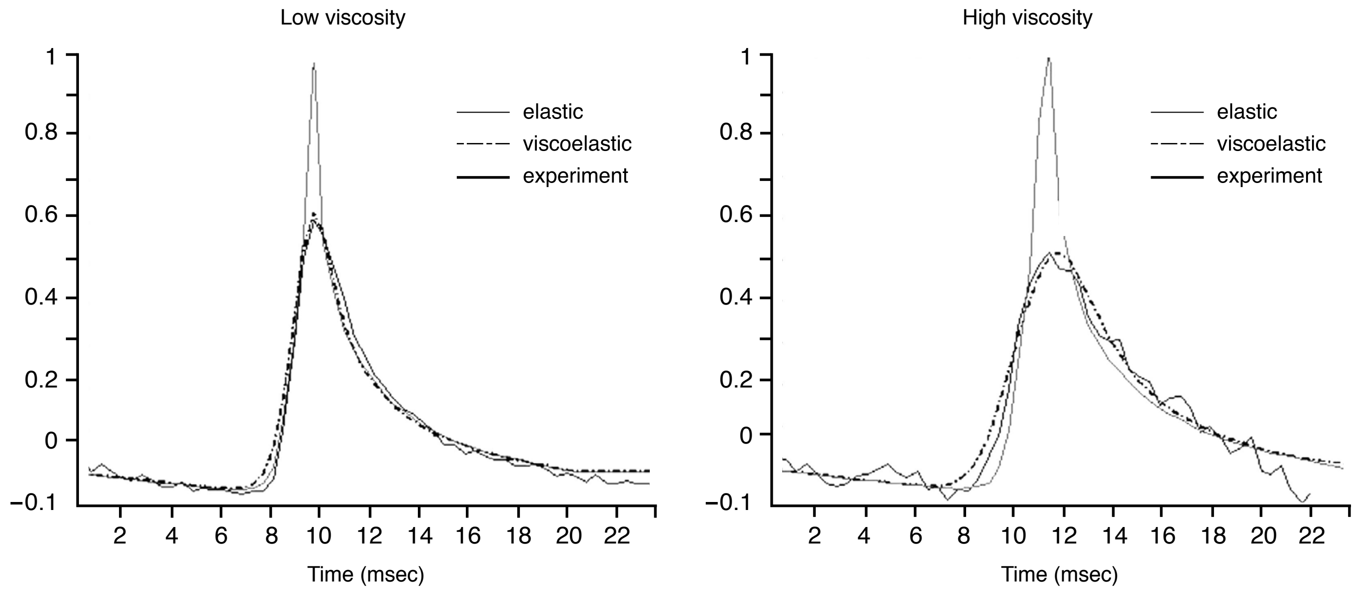

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Fig. 4. **Influence of viscosity on the shear wave temporal shape for a low-viscosity phantom (A) and a high-viscosity phantom (B).** Reprinted from Bercoff et al. IEEE Trans Ultrason Ferroelectr Freq Control 2004;51:1523-1536 [22], with permission from the IEEE Xplore Digital Library.

The Young Modulus, Shear Modulus, Shear Wave Speed, and Stiffness

Various elastic moduli are used to define the tissue elasticity [6,25]. The elasticity of soft tissue is most commonly reported using the Young modulus (*E*), which is the resistance of a material to deformation in uniaxial compression or tension (kPa) [6]. The shear modulus (μ) is the resistance to shear force (kPa) [26]. In soft tissue, two modes of wave propagation occur: longitudinal waves (*cL*), in which the particles oscillate in the direction of wave propagation; and transverse waves (*cT*), where the particles oscillate in the direction transverse to the wave propagation. The transverse wave propagation speed is called the shear wave speed (or shear speed, m/sec), and is one of the terms of elastic moduli [6,16]. The Young modulus is defined as $E=3\mu=3\rho cT^{2}$, where ρ denotes the tissue density [25]. As a result, in tissue with a density not equal to 1.0 g/mm^3, the shear speed is not exactly correlated with the Young modulus. The United States Food and Drug Administration has approved the use of shear speed (m/sec). Most commercially available US systems are able to display the Young modulus (kPa) [8].

Stiffness is a somewhat different concept from the elastic modulus. It is defined as force over displacement and has units of force per distance. It is a measure of the rigidity of an object, and it is influenced by the elastic modulus of the object itself, shape, and size [27].

Intersystem Variability in Elasticity

Statistically significant variability has been observed in elasticity quantification according to the vendor system and the depth of the measurements [28,29]. Therefore, to ensure reproducibility of the measurements, a patient should be followed using the same US transducer from the same machine, and the threshold values suggested in previous research should not be applied to other systems [28]. For meaningful comparisons of the absolute elasticity of the lesions and the serial assessment of disease states, the sources of variability should first be characterized [27]. The impacts of viscosity and dispersion may be sources of intersystem variability [29].

Clinical Applications of Shear Wave US Elastography to Tendon and Muscle

Tendons

Imaging of tendons with US elastography is not easy, and the relevant techniques are still being developed. In recent years, US elastography has been actively applied to tendon imaging in normal healthy volunteers and in individuals with abnormal tendons [30]. Tendon is a very hard tissue in its normal state, and in case of degeneration or injuries, its stiffness can change to various degrees (Fig. 5) [31].

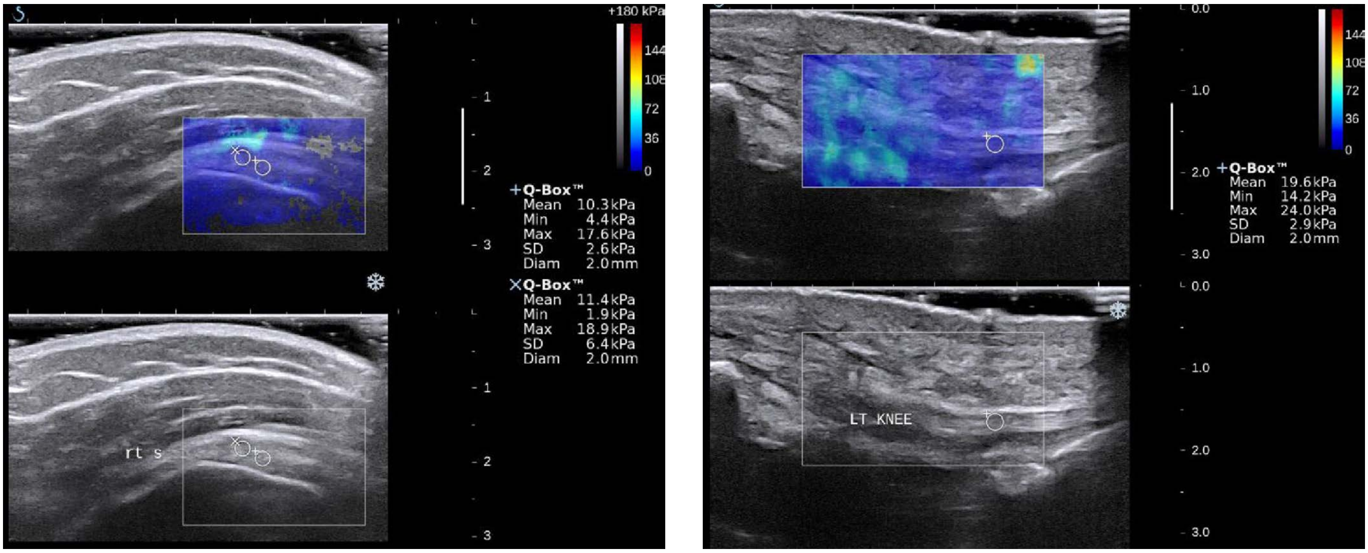

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Fig. 5. **Shear wave elastography of the tendons.**
A. The right supraspinatus tendon of a 36-year-old woman shows heterogeneously increased echogenicity in a grayscale image (bottom), suggestive of tendinosis. In shear wave elastography (top), it shows values of 10.3–11.4 kPa, which are relatively low values for the Young modulus. B. The left patellar tendon of an 81-year-old man appears normal in a grayscale image (bottom). In shear wave elastography (top), it shows a homogeneously colored elastogram, and a relatively low value of the Young modulus, measured at 19.6 kPa.

The Young modulus of tendon is approximately 400–1,300 kPa, and in a normal volunteer study using shear wave elastography, the elasticity of the Achilles tendon was measured as 6–242 kPa in the neutral position [21], which means the shear wavelengths can be greater than tendon thickness. As a result, guided wave propagation may lead to a false evaluation of the Young modulus, which is the reason that shear wave velocities are usually preferred to Young moduli in tendon imaging [6,22,32]. In addition, the high shear wave velocity of hard tendon can exceed the upper limit of measurement of shear wave velocity of the device (maximum scale value) [32,33]. In the Acuson S3000 ultrasound system Virtual Touch imaging quantification (VTIQ; Siemens Medical Solutions, Malvern, PA, USA), the maximum scale value is 10 m/sec [33], and in the Imagine Aixplorer system (Supersonic Imagine, Aix-en-Provence, France), it is 16.3 m/sec [34].

Due to the high anisotropy, tendon imaging requires the US beam position to be perfectly parallel or perpendicular to the tendon fibers; however, the tendons may have a complex structure due to interpenetration and rotation of fibers from muscles, and sometimes proper positioning of the US beam cannot be achieved [32,33]. Several studies of the Achilles tendon have reported large differences in elasticity values in the transverse (axial) and longitudinal (sagittal) orientations of the probe and the neutral position, extension, and dorsiflexion of the ankle [32,33,35,36].

In a recent study, the shear wave velocity of a normal Achilles tendon in the neutral position was reported to be 15.55 m/sec in the sagittal orientation of measurement and 5.29 m/sec in the axial orientation, with an anisotropic coefficient of 0.66; while ankle plantar flexion was 7.03 m/sec in the sagittal orientation and 4.76 m/sec in the axial orientation, with an anisotropic coefficient of 0.33 [32].

In another study, in the relaxed position, the shear wave velocity of a normal Achilles tendon was 8.26 m/sec in the sagittal orientation, and 4.10 m/sec in the axial orientation, with an anisotropic coefficient of 0.50 [33]. The sagittal orientation and the extended tendon position showed markedly higher shear wave velocities in several studies in both normal and abnormal Achilles tendons, which may explain the low intraobserver-interobserver reproducibility and repeatability in many studies [35,36]. However, when an ankle fixator was used to standardize the position of the feet and the degree of flexion of ankle and a standardized measurement site was used for the measurement of the Achilles tendon throughout the scanning process, the reproducibility improved [37,38].

Tendinopathy is defined as the various painful conditions occurring from mechanical, degenerative, and overuse diseases, and it is associated with degeneration and disorganization of the collagenous structure, changes in the proteoglycan and water contents, increased cellularity, fatty infiltration, and neovascularization [31].

In one study, the mean elasticity value for normal Achilles tendons was approximately 291 kPa (range, 261 to 300 kPa) and that of ruptured Achilles tendons was approximately 56 kPa (range, 3 to 228 kPa) in the neutral position [39]. The elastogram map of the normal tendon showed that it was hard and homogeneous, and the ruptured tendon was heterogeneous [39].

The stiffness of the Achilles tendon increases after both static stretching and long-term exercise, especially in the nondominant leg, in frequent exercisers [37,38]. This suggests that shear wave elastography can be used to evaluate and follow the effects of exercise in healthy athletes and patients undergoing rehabilitation treatment.

In another study of patients who had undergone surgical repair of a torn Achilles tendon, the repaired tendon elasticity showed a positive correlation with the functional outcome of the tendon during a 48-week follow-up period [40], suggesting that shear wave elastography can be used to evaluate biomechanical information regarding the healing process in Achilles tendons and to predict tendon function (Table 1).

Studies have been carried out of the patellar tendon and other tendons. In Achilles, patellar, and epicondylar tendinopathies, a decrease in tendon stiffness values was correlated with the patients' symptom scores, demonstrating the promise of shear wave elastography during follow-up for tendinopathies. Furthermore, the diagnostic accuracy of tendinopathy has also improved with shear wave elastography [41]. In older patients (61–70 years), the shear wave velocity of the patellar tendon was found to be significantly lower than that of other age groups [42]. The correlation between aging and Achilles tendon stiffness remains controversial.

Muscles

The use of shear wave elastography to study muscles has increased exponentially in the last few years. Because it allows real-time visualization of muscle stiffness during active or passive muscle movement, shear wave elastography has become the most promising modality for evaluating muscle [43,44]. In clinical practice, muscle spasticity in stroke, spinal cord injury, and myopathy has been evaluated through manual palpation, but only in a qualitative and subjective manner. The same has taken place for myofascial pain and in patients undergoing muscle rehabilitation exercises [43].

Dynamic shear wave elastography can dynamically quantify and evaluate isolated individual muscle elasticity during relaxation and contraction (Fig. 6). Muscle stiffness is a very important primary impairment, especially in children with spastic cerebral palsy, because it causes joint motion limitation and fixed contractures.

Shear wave elastography can be used to evaluate and compare the muscles of bilateral limbs, and yield an improved understanding of an individual patient without the need for invasive muscle biopsy or complex lab-based dynamometry, as an extension of the physical exam [45,46].

Studies have sought to develop a reliable protocol for shear wave elastography of muscles to overcome the difficulties in the evaluation of muscles, including voluntary contraction, fatigue and trembling, muscle contraction caused by discomfort, and fatty infiltration of muscles, as well as innately high anisotropy, the variety of positions, and degree of contraction [47].

Various studies have been conducted on muscles. The normal elasticity of muscle during relaxation and contraction has been measured: the values for the tibialis anterior muscle were 40.6 kPa and 268 kPa, those of the gastrocnemius muscle were 16.5 kPa and 225 kPa, and those of the soleus muscle were 14.5 kPa and 55 kPa, respectively [43]. These results were correlated with magnetic resonance elastography, which reported values of 5–40 kPa in the resting state, and up to 300 kPa during contraction [48].

The normal value for the shear wave velocity of the supraspinatus

Table 1. Summary of the normal and abnormal ranges of shear wave speed in the Achilles tendon according to the literature

Status	Shear wave speed/Young modulus	Position	Measurement direction	Anisotropic coefficient	Reference
Normal	74.4±45.7 (6–242) kPa	Neutral	Longitudinal	–	[21]
	51.5±25.1 (10–111) kPa	Neutral	Transverse		
Normal	15.55 m/sec	Neutral	Longitudinal	0.66	[32]
	5.29 m/sec	Neutral	Transverse		
	7.03 m/sec	Plantar flexion	Longitudinal	0.33	
	4.76 m/sec	Plantar flexion	Transverse		
Normal	8.26 m/sec	Relaxed	Longitudinal	0.50	[33]
	4.10 m/sec	Relaxed	Transverse		
Normal or abnormal	Markedly higher	Extended	Longitudinal	–	[34,35]
Normal	291 (261–300) kPa	Neutral	–	–	[39]
Ruptured	56 (3–228) kPa				

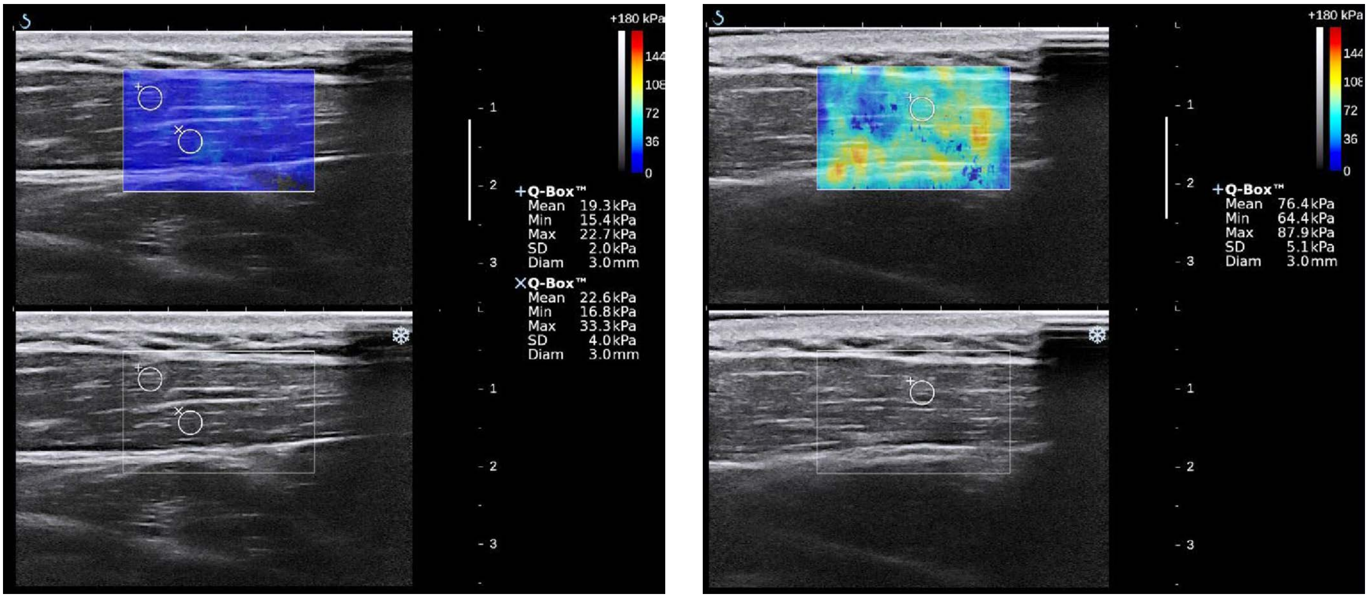

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Fig. 6. Shear wave elastography of the deltoid muscle of a 49-year-old woman.
A. In the resting state, a homogeneous blue color elastogram is noted in the Q-box. The elasticity was measured at 19.3–22.6 kPa. B. During strain, the deltoid muscle shows markedly increased elasticity, measured at 76.4 kPa, and relatively homogeneous yellow in the color elastogram.

muscle was measured to be 3.0 m/sec [49]. Among the hamstring muscles, the shear modulus of the semimembranosus was significantly higher than that of other muscles [50]. A study on the passive stretching of a healthy tibialis anterior muscle reported that the Young modulus increased exponentially from 7 to 35 kPa, according to the increase in the plantar flexion angle of the ankle [44]. Elongation of the deltoid muscle by an external fixator showed an exponential increase in the Young modulus during passive stretching [51].

The rectus femoris muscle and lateral head of the gastrocnemius muscle have been reported to have a significantly higher Young modulus in younger individuals than in the elderly. No significant difference was found in the soleus muscle between these groups [52].

The supraspinatus muscle showed a shear wave velocity that tended to decrease with fatty infiltration at Goutallier stage III; however, it increased in stage IV [49].

The intraobserver and interobserver reproducibility of shear wave elastography in muscle studies ranged from poor to excellent, but the majority of findings were fair or good [53,54].

In a comparative study of GNE myopathy patients and a healthy group, the shear wave velocity of the muscles was significantly higher in measurements with a transverse orientation and in the deep muscles than in the longitudinal orientation and the superficial muscles. GNE-myopathy patients had significantly lower mean values of shear wave velocity than the normal group, and the serial follow-up evaluation could be used to assess changes in muscle volume and strength in the myopathy group [55].

A study of the effects of prolonged exercise on muscle stiffness showed a significant decrease in the quadriceps shear modulus after 48 hours after exercise compared to baseline [56], likely due to the development of inflammation and muscle swelling. The results suggest the possibility of monitoring physiological and pathological changes in muscles.

In chronic stroke patients, several patterns of shear moduli and torque response associated with passive elbow extension were noted in a recent study [57], and it was simultaneously suggested that wide individual variation was present in passive movement in chronic stroke patients. It was additionally proposed that shear wave elastography has potential as an alternative or complement to electromyographic evaluation [57]. In a study of Parkinson disease patients and a healthy control group, the biceps brachii muscle was measured; symptomatic arms showed a Young modulus of 50.87 kPa (median value); asymptomatic arms, 40.06 kPa; and healthy controls, 23.70 kPa [58].

In children with spastic cerebral palsy, the passive muscle stiffness of the lateral gastrocnemius muscle at different degrees of plantar flexion was investigated [46]. The shear modulus of the muscle was

Table 2. Summary of the normal and abnormal ranges of shear wave speed of various muscles according to the literature

Status	Muscle	Shear wave speed/Young modulus	Position	Reference
Normal	Tibialis anterior	40.6 kPa	Relaxed	[43]
		268 kPa	Contraction	
	Gastrocnemius	16.5 kPa	Relaxed	
		225 kPa	Contraction	
	Soleus	14.5 kPa	Relaxed	
		55 kPa	Contraction	
Normal	Tibialis anterior	7 kPa	Resting	[44]
		35 kPa	Passive stretching	
Normal	Supraspinatus	3.0 m/sec	–	[49]
Parkinson disease	Biceps brachii	50.8 kPa	Symptomatic arm	[58]
		40.0 kPa	Asymptomatic arm	
		23.7 kPa	Healthy control	
Children with cerebral palsy	Gastrocnemius lateral head	15–25 kPa	–	[46]

higher in children with spastic cerebral palsy than in unaffected children, and ranged from 15 to 25 kPa [46].

In medial tibial stress syndrome patients, the shear moduli of the medial head of the gastrocnemius, the lateral head of the gastrocnemius, and the soleus, plantaris, and tibialis anterior muscles were significantly higher than the corresponding values in the normal control group (Table 2) [59].

Recent studies of patients with lower back pain undergoing rehabilitation exercises for trunk stability reported that trunk exercise could improve treatment outcomes because the transversus abdominis muscle contributes to the control of spinal motion. Shear wave elastography could provide real-time feedback during trunk exercise, and more selective measurements of muscles, especially deep muscle, are possible without cross-talk contamination of the adjacent muscles and with reasonable reliability and reproducibility [60].

In an experimental study conducted using calves from cadavers, the muscle shear modulus was significantly reduced by 50% after removing the covering skin tissues [61], which means that the skin maintains the mechanical properties of the muscle. It has also been reported that the underlying bone had an influence on the shear wave velocity of the overlying muscles [62]. The shear wave velocity decreased significantly with increasing depth and when there was underlying bone below the ROI [62].

The Radiological Society of North America Quantitative Imaging Biomarkers Alliance for Shear Wave US Elastography

During the past two decades, shear wave elastography has been used in medicine, and several physical misconceptions and nomenclatorial inaccuracies have been noted [27]. For example, Young modulus values are expressed as kilopascals (kPa) and shear wave velocity is expressed in units of speed (m/sec). To interpret different studies, the most appropriate result may be shear wave velocity [27]. When we use the formula of the Young modulus ($E=3\mu=3\rho cT^2$), ρ is the tissue density and it is assumed to be constant at 1,000 kg/m^3, under the assumption of a purely elastic model, which is valid for isotropic tissues such as the liver or the thyroid gland [25]. However, muscle and tendon are not isotropic or purely elastic, and we should calculate both the elasticity and the viscosity of tendons [32].

Information regarding frequency is also important, because higher-frequency shear waves travel faster, and the viscosity and geometry of soft tissue can cause the shear wave speed to vary with frequency [27]. There are still several questions to resolve regarding shear wave elastography.

In 2008, the Radiological Society of North America created the Quantitative Imaging Biomarkers Alliance (QIBA) with United States federal government representatives (the Food and Drug Administration, National Institutes of Health, and National Institute of Standards and Technology, among others), imaging system manufacturers, pharmaceutical companies, academics, and clinicians to establish guidelines regarding quantitative imaging and the use of imaging biomarkers in clinical practice [29,63–65].

They used standardized ultrasound phantoms and standardized methods to evaluate and understand the sources of bias and variance in shear wave elastography [29,64–66]. In recent phantom studies, researchers found statistically significant differences in shear wave speed estimation among systems and according to the

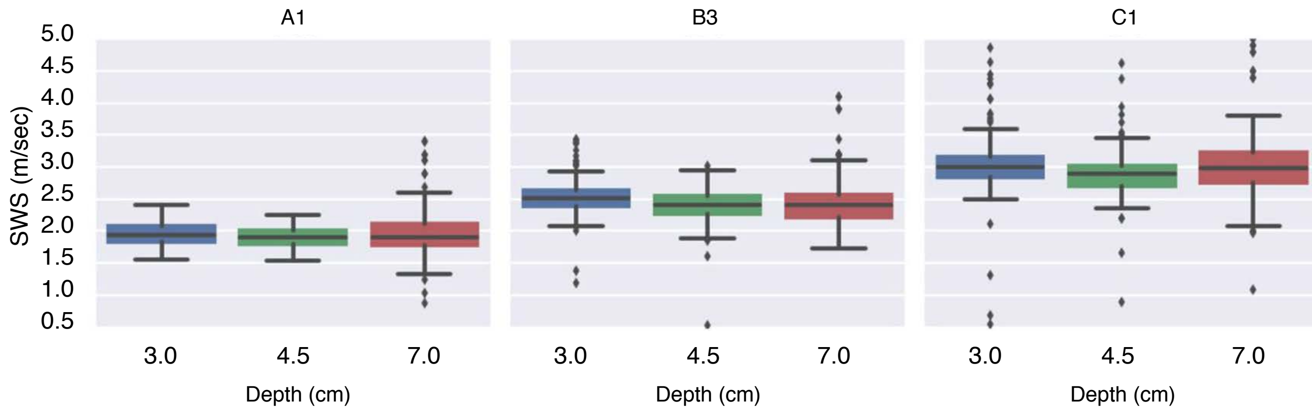

Fig. 7. **Shear wave ultrasound measurements of multiple centers using multiple vendor systems as a function of phantom and focal depth.** The phantoms represent: healthy liver (A1), mildly fibrotic (B3), and significantly fibrotic (C1) tissue. Shear wave speed (SWS) measurement variance increased as a function of higher stiffness (A1<B1<C1) and focal depth (3 cm, 4.5 cm, and 7 cm), which is consistent with the limitations of finite shear wave spatial and temporal sampling, and a decreased signal-to-noise ratio with increasing stiffness and depth. Despite some intersystem variability, current-generation ultrasound SWS imaging systems are able to differentiate viscoelastic material properties (P<0.01, one-way analysis of variance) spanning healthy to fibrotic liver. Reprinted from Palmeri et al. (2016) RSNA QIBA Ultrasound shear wave speed phase II phantom study in viscoelastic media, with permission from the IEEE Xplore Digital Library [29].

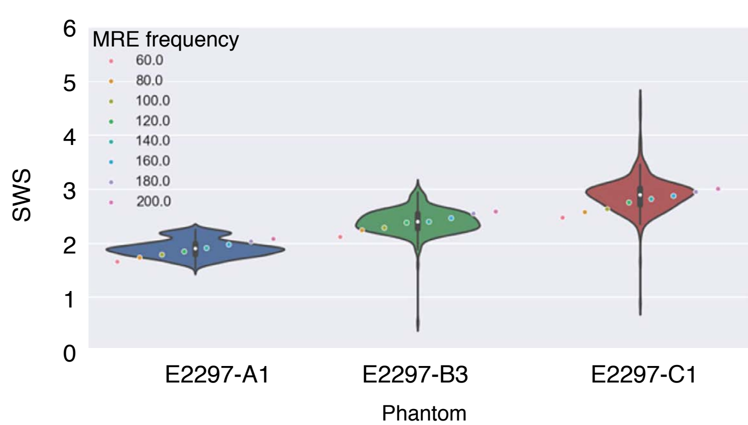

Fig. 8. **Violin plots (blue, green, and red) of the distributions of 24 ultrasound (US) elastography shear-wave speed (SWS) measurements for each of the three phantoms with the nine ultrasound systems.** The multi-color dots show the magnetic resonance elastography (MRE) SWS measurements at eight different shear wave frequencies. The SWS measured by MRE show a linear dependence on shear-wave frequency for all three phantoms. US elastography SWS measurements corresponded most closely to MRE SWS measurements at 140 Hz. Reprinted from Palmeri et al. (2016) RSNA/QIBA Ultrasound Shear Wave Speed Biomarker Committee [69], with permission from the Palmeri et al. and Radiological Society of North America.

depth of the object (Fig. 7); the intersystem variability was usually less than 3% of the shear wave velocity for the typical liver imaging parameters, and in the viscoelastic media it was less than 17.7% [29,66,67]. Several sources of system-dependent bias exist, such as arrival time estimation noise, speckle bias, hardware fluctuations, phase aberration, pulse repetition frequency errors, beamforming errors, and mismatches between the coupling medium and the speed of sound [66].

More recently, the shear wave US viscoelastography method has been developed [67,68]. It can be used to quantify shear wave velocity more reliably, considering both the shear storage modulus and the loss modulus in frequency-dependent viscoelastic tissue [67,68].

Researchers also found that US shear wave velocity corresponded to measurements made using magnetic resonance elastography, especially at 140 Hz (Fig. 8) [69]. The shear wave showed temperature dependence in the viscoelastic phantom, but at room temperature this effect was not significant; and at higher US attenuation, the shear wave velocity showed more biased values [69].

Conclusion

Shear wave elastography is a promising diagnostic modality for the musculoskeletal system, and the accurate measurement of muscle and tendon elasticity has a major impact in clinical practice.

The accuracy of measurements is critical for the use of shear wave elastography to diagnose patients, and we should develop a complete understanding of the fundamental properties of this measurement technique.

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Conflict of Interest
No potential conflict of interest relevant to this article was reported.

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