

[CLINICAL COMMENTARY]

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Rehabilitative Ultrasound Imaging: Understanding the Technology and Its Applications

From an historical perspective, ultrasound imaging (USI) has been used for medical purposes since the 1950s. The primary use of USI continues to be for traditional radiological goals, which consider the morphological characteristics and structural integrity of various organs and tissues. However, as the technology has been embraced as a safe, portable, objective, and relatively inexpensive means of examination, the ingenuity and diversity of applications has extended beyond these realms.

Ultrasound imaging related to musculoskeletal rehabilitation has been developing rapidly since the 1980s. The first report of muscle imaging linked to rehabilitation was in 1968, when Ikai and Fukunaga⁶³ related the size of the upper arm muscles to strength. However, it was the work of Dr Archie Young and colleagues

at the University of Oxford in the 1980s that sowed the seeds for the use of USI by physical therapists. A striking finding of their work was how dramatic limb muscle wasting is underestimated with a tape measure.¹⁴⁴ Several studies of the quadriceps muscle followed, including investigation of the effect of knee joint injury,

strength-training protocols, and aging on muscle size, and the relationship between muscle size and strength in different populations (see Stokes and Young¹²² for a review). This early research used compound B-scanning, which enabled whole cross sections of large muscles to be captured, because the image could be built up as the transducer was moved over the skin. The compound technique, which was expensive, was phased out as a routine tool and replaced by real-time USI (definitions are provided in the **APPENDIX**) both in general medical and musculoskeletal settings.

A recent (1990s) resurgence in the interest of rehabilitative applications of USI has been seen amongst clinical therapists. This stems from a series of studies in which USI was used to detect atrophy of the lumbar multifidus (isolated to the side and spinal level of symptoms) in individuals with acute low back pain (LBP),⁵¹ as well as to determine that recovery of this muscle was not automatic when pain subsided,⁴⁹ thus required specific training to reduce risk of future episodes.⁴⁷ In addition, these studies

● **SYNOPSIS:** The use of ultrasound imaging by physical therapists is growing in popularity. This commentary has 2 aims. The first is to introduce the concept of rehabilitative ultrasound imaging (RUSI), provide a definition of the scope of this emerging tool in regard to the physical therapy profession, and describe how this relates to the larger field of medical ultrasound imaging. The second aim is to provide an overview of basic ultra-

sound imaging and instrumentation principles, including an understanding of the various modes and applications of the technology with respect to neuromusculoskeletal rehabilitation and in relation to other common imaging modalities. *J Orthop Sports Phys Ther* 2007;37(8):434-449. doi:10.2519/jospt.2007.2350

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suggested that the biofeedback provided by USI might facilitate the relearning process. Since that time, applications of USI with respect to many other muscles of the trunk and limbs continue to be investigated.¹¹⁹

Current applications of USI in rehabilitation essentially fall into 2 distinct areas of musculoskeletal imaging: rehabilitative USI (RUSI) and diagnostic imaging. The former, which is the topic of this special issue, includes evaluation of muscle structure (morphology) and behavior, as well as the use of USI as a biofeedback mechanism. Specifically, this includes the measurement of morphological features (morphometry), such as muscle length, depth, diameter, cross-sectional area, volume, and pennation angles; changes in these features and the impact on associated structures (fascia and organs such as the bladder) with contraction; tissue movement and deformation (eg, high-frame-rate USI and elastography); and qualitative evaluation of muscle tissue density. Alternatively, diagnostic USI involves examining the effects of injury or disease on ligament, tendon, and muscle tissues, which requires different skills and training than those needed for RUSI.¹⁴

In May 2006, the first international meeting on RUSI was hosted by the US Army-Baylor University Doctoral Program in Physical Therapy in San Antonio, TX. The purpose of the symposium was to develop best practice guidelines for the use of USI for the abdominal, pelvic, and paraspinal muscles, and to develop an international and collaborative research agenda related to the use of USI by physical therapists. At that symposium the participants agreed on the use of the term *RUSI*. In addition, a position statement (below) was created to help define this emerging tool in the field of physical therapy.¹²⁸ This statement, along with a visual representation of how the practice of RUSI fits into the larger field of medical USI, was endorsed by delegates (**FIGURE 1**):

“RUSI is a procedure used by physical therapists to evaluate muscle and related

soft tissue morphology and function during exercise and physical tasks. RUSI is used to assist in the application of therapeutic interventions aimed at improving neuromuscular function. This includes providing feedback to the patient and physical therapist to improve clinical outcomes. Additionally, RUSI is used in basic, applied, and clinical rehabilitative research to inform clinical practice. Currently, the international community is developing education and safety guidelines in accordance with World Federation for Ultrasound in Medicine and Biology (WFUMB). Dated: 10 May, 2006.^{7,128}

In addition to defining the scope of USI with respect to physical therapy, the position statement and diagram are intended to guide therapists in acknowledging professional boundaries, as ultimately the delegates’ goal is to see RUSI accepted within the medical-imaging field. However, as the use of USI (both rehabilitative and diagnostic) by physical therapists is in its infancy, the need to establish training facilities for therapists in conjunction with other imaging disciplines, including their professional bodies, where possible, is recognized as a priority.

This commentary aims to provide an overview of basic USI and instrumentation principles as they relate to RUSI.

This will include an introduction to the various modes of imaging, how USI fits with respect to other more commonly known imaging technologies, the type of information that USI applications can provide, how these applications may be of value to the researcher and clinician, as well as potential future lines of investigation.

BASIC PRINCIPLES OF SOUND WAVE PROPAGATION AND INSTRUMENTATION

THIS SECTION IS INTENDED TO PROVIDE a basic understanding of the principles that underlie USI. The generic characteristics of USI units and the physical properties of sound wave propagation will be discussed. As a complete appraisal of these topics is not possible in this forum, the reader is encouraged to refer to more thorough resources for further discussion.^{73,138}

The Physical Properties of Sound

Ultrasound is defined as sound with a frequency greater than 20 000 Hz, which is the upper limit of the range registered by the human ear. USI uses sound waves primarily in the range of 3.5 to 15 MHz. Ultrasound waves behave according to

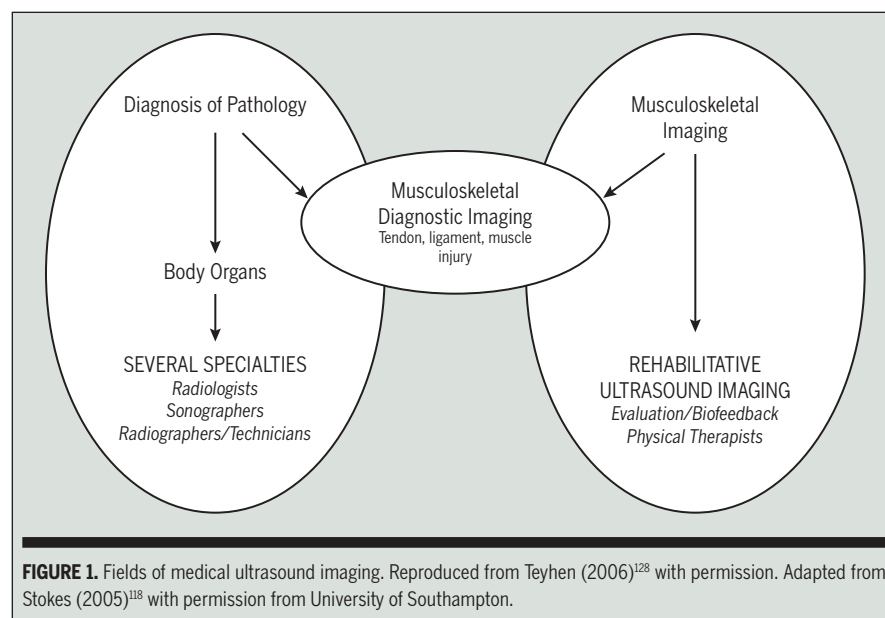


FIGURE 1. Fields of medical ultrasound imaging. Reproduced from Teyhen (2006)¹²⁸ with permission. Adapted from Stokes (2005)¹¹⁸ with permission from University of Southampton.

principles that apply to all sound waves, which at the most fundamental level are mechanical waves that travel via particle vibration. Specifically, the source of a sound creates oscillatory vibrations that affect particles in the medium that lies adjacent to it. These particles, in turn, affect their adjacent particles, and so on. This process is referred to as wave propagation.⁸¹ How far a sound wave propagates and whether an echo is produced depends on the strength of the sound source, the properties of the media through which the sound has to travel, and the number, shape, and properties of the objects it encounters.⁷³ These behaviors can be summarized by the principles of penetration and attenuation.

Penetration Penetration refers to the ability of sound to travel (depth) and is influenced by the intensity (strength or loudness), frequency, and speed of a sound wave. The intensity of an ultrasound wave refers to the rate at which energy is delivered per unit area and is determined by the total power output (W) of an ultrasound transducer, divided by its area (cm²), and is expressed in units of mW/cm². As the intensity of an ultrasound wave increases, so does the depth it can penetrate, the strength of the echo that it can generate, and the potential it has to induce biological effects (heat and cavitation) within the tissues it is traveling through.

Frequency is defined as the number of oscillations that a wave undergoes in 1 second and is expressed in hertz (Hz). The higher the frequency of sound, the less the emerging wave will diverge. Due to their relatively high frequency, ultrasound waves are cohesive and can be used to selectively expose a target area. The frequency of an ultrasound wave is determined in the construction of the transducer assembly. As a general rule, the lower the frequency of a sound wave the farther it will penetrate.

The speed at which an ultrasound wave travels is determined by the density and stiffness of the structure or medium it is traversing. The more rigid the media the faster sound will travel through it.

The average speed at which sound travels through soft tissue is 1540 m/s, which is similar to the velocity that it would travel through water (1485-1526 m/s).⁸⁵ Fat is less stiff than most soft tissue. Hence sound traverses it at a slightly slower speed (1450 m/s). Muscle and bone are stiffer and consequently sound propagates faster through them (1585 m/s and 3500m/s, respectively).

Attenuation As an ultrasound wave propagates, it encounters changes in tissue densities, or interfaces. Each tissue or medium has a characteristic resistance to sound referred to as acoustic impedance. This value is dependent upon the density of the medium and the speed at which sound can travel through it. At each interface between media of dissimilar impedance, an ultrasound wave will react and lose energy. Consequently, the energy within a sound wave decreases as it penetrates, until it is completely dispersed. This phenomenon, referred to as attenuation, occurs through the processes of reflection, scattering, refraction, and absorption. Although the first 3 processes contribute to the dispersion of an ultrasound wave, most of its energy is absorbed by the surrounding tissue in the form of heat.⁷³

When a sound wave encounters an interface, the portion that is reflected back to its source is referred to as “reflection” and serves as the basis for image formation. The strength of a reflection depends on the size of the reflecting medium, the roughness of its surface, the incident angle of the sound wave, and the difference in impedance of the 2 media that create the interface.⁷³ The more regular the surface, the greater the difference in impedance between tissues and the more perpendicular the incidence angle; hence the greater the reflection and brighter (more white) the interface appears within the ultrasound image. An obvious example of this is the interface between bone and muscle (**FIGURE 2**). Not only is there a significant difference in the impedance of these 2 tissues, but bone attenuates a high percentage of the incident sound

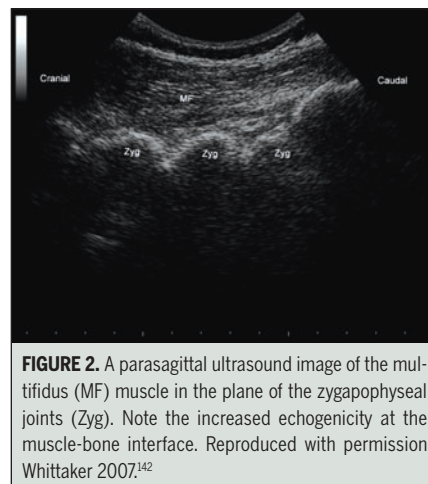


FIGURE 2. A parasagittal ultrasound image of the multifidus (MF) muscle in the plane of the zygapophyseal joints (Zyg). Note the increased echogenicity at the muscle-bone interface. Reproduced with permission Whittaker 2007.¹⁴²

wave and, consequently, obscures the view of deeper structures.

A sound wave can also scatter or refract when it encounters an interface between heterogeneous media. If, for instance, the structures that comprise an interface are very small, portions of the wave will be scattered. Those portions that travel back to the transducer are used in image formation, while those that scatter (the majority) are not. Alternatively, if there is either a significant difference in the speed that sound can travel through the 2 tissues or if the interface is not at a right angle to the ultrasound wave, the wave will change its direction when it crosses the boundary. This is referred to as refraction and it too can be a detriment to image formation through production of positional errors.

The practical implication of attenuation is that it limits penetration and consequently the depth of the images that can be generated.⁷³ Attenuation and frequency have a direct relationship: the higher the frequency of an ultrasound wave, the greater the attenuation and the more shallow its penetration. Conversely, the more attenuation, the more reflection and the better the detail resolution (ability to show detail) demonstrated in the ultrasound image. Consequently, the choice of frequency used for an imaging application will be dependent upon the depth of the region or structures of interest. Higher frequencies (7.5-10.0 MHz) are more valuable for ex-

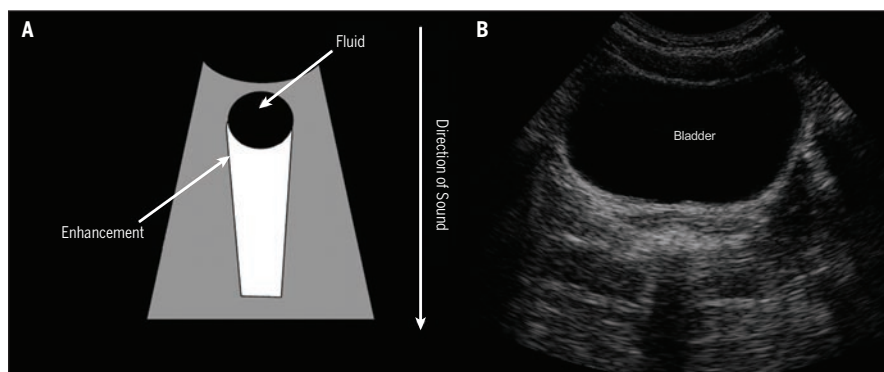


FIGURE 3. (A) Depiction of the enhancement of a region deep to a fluid-filled structure. Enhancement occurs as there is less attenuation of the propagating sound wave as it travels through a fluid-filled structure. (B) A transverse ultrasound image demonstrating enhancement of the midline pelvic floor structures deep to the bladder. Reproduced with permission Whittaker 2007.¹⁴²

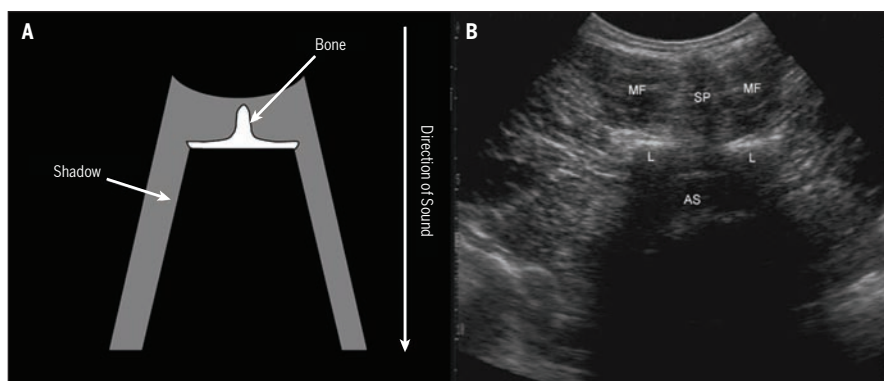


FIGURE 4. (A) Depiction of how an acoustic shadow forms behind a strongly attenuating (hyperechoic) structure such as bone. (B) A transverse ultrasound image demonstrating acoustic shadowing (AS) caused by the posterior elements of a lumbar vertebra. Abbreviations: L, lamina; MF, multifidus; SP, spinous process. Reproduced with permission Whittaker 2007.¹⁴²

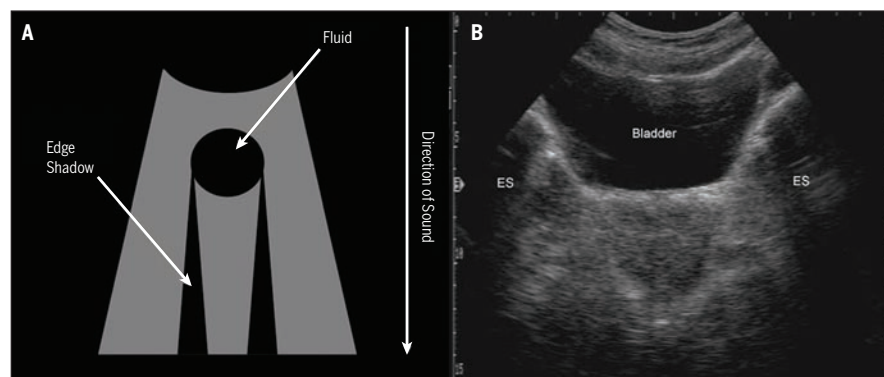


FIGURE 5. (A) Depiction of an edge shadow produced when a sound wave is refracted (bent) around the edges of a fluid filled structure. (B) A transverse ultrasound image demonstrating edge shadowing (ES) caused by the bladder. Reproduced with permission Whittaker 2007.¹⁴²

frequency transducer that can image an area of interest should be used.¹³⁸

Artifact USI devices generate images based upon several assumptions: sound travels in straight lines, echoes only originate from objects located in the 2 dimensions of the sound beam, the amplitude of an echo is directly related to the reflecting or scattering properties of the objects it encounters, and the speed at which sound travels through all the tissues is a constant 1540 m/s.⁷³ If any of these assumptions are violated, incorrect representations of anatomy can occur. These incorrect representations are referred to as “artifacts” and can also be the result of improper equipment operation or imaging technique. Artifacts can be both a help and a hindrance, and result in situations in which structures are either not real, missing, improperly located, or of improper brightness, shape, or size. Those which have a greatest impact on RUSI include enhancement, shadowing, and reverberation.

Acoustic enhancement refers to an increase in the amplitude of the ultrasound echo coming from a structure that lies behind a weakly attenuating structure, such as a fluid-filled cavity (eg, bladder [FIGURE 3]).⁷³ As the ultrasound device assumes that there is uniform attenuation of the ultrasound wave as it propagates, the tissues on the far side of the transmitting structure appear brighter than they should, as they are being exposed with a less attenuated beam. When this occurs, gain settings can be manipulated to compensate.

Acoustic shadowing is the opposite of enhancement. It refers to a reduction in the sound wave echo from structures that lie behind a strongly attenuating structure, such as bone (FIGURE 4).¹¹⁰ Specifically, ultrasound waves hit something that blocks their path and everything behind the blocking structure appears black, as if it were within an “acoustic shadow.” A shadow can also occur as a sound beam is refracted (bent) from its original path by passing close to a large, curved, fluid-filled structure (FIGURE 5).^{43,61}

amining superficial structures (superficial muscles, ligament, and tendons) and lower frequencies (3.5-5.0 MHz) for deeper

structures (deeper muscles, the bladder, and contents of the abdominal/pelvic cavities). As a general rule, the highest

Reverberation refers to multiple reflections and is a result of ultrasound echoes bouncing between tissue layers and the transducer. Specifically, when an echo from a highly reflective surface that lies parallel to the transducer face returns to the transducer, a portion may be reflected back into the tissue to meet the same interface, where it again is reflected back to the transducer. Due to the time delay of the echoes being registered at the transducer, the depth of that interface is portrayed progressively deeper within the tissue. As the reflective echoes become weaker, the artifact fades out (FIGURE 6).

Instrumentation

A typical USI device is a pulsed-echo (generates a series of short ultrasound waves at regular intervals) instrument consisting of 2 components: a transducer assembly (commonly referred to as a “transducer” or “probe”), and an imaging system. The transducer is responsible for generating ultrasound waves, as well as receiving the ultrasound echoes returning from the tissues and converting them into electrical signals. The imaging system is the component of the technology that receives the electrical signals representing the echo from the transducer and processes them so that they can be displayed as a digital image.

Imaging System A USI system consists of 4 generic components: the beam former, signal processor, image processor, and visual display.⁷³ In general terms, the beam former is responsible for generating the electrical impulses that drive the transducer assembly, as well as for amplifying and digitizing the electrical signal returning from the transducer assembly that represents the ultrasound echo. The signal processor is responsible for filtering and compressing the electrical signal before the image processor converts the signal into an image presented on the instruments display.⁷³

Transducer Assembly (Probe) A transducer assembly houses an array of crystals (transducers), their electrical connections, an acoustic lens, and damping material.

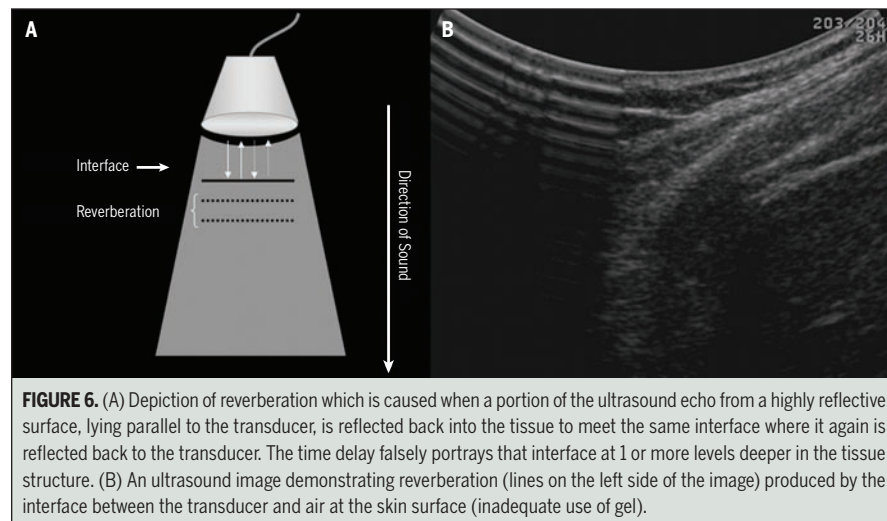


FIGURE 6. (A) Depiction of reverberation which is caused when a portion of the ultrasound echo from a highly reflective surface, lying parallel to the transducer, is reflected back into the tissue to meet the same interface where it again is reflected back to the transducer. The time delay falsely portrays that interface at 1 or more levels deeper in the tissue structure. (B) An ultrasound image demonstrating reverberation (lines on the left side of the image) produced by the interface between the transducer and air at the skin surface (inadequate use of gel).

By definition, a transducer is a device that converts one form of energy to another.⁷³ Ultrasound transducers (also referred to as “elements” or “crystals”), commonly a ceramic formulation of lead zirconate titanate, are piezoelectric elements that produce voltage (electrical energy) when deformed by an applied pressure such as a sound wave (acoustic energy).⁷³ Although not technically accurate, a transducer assembly is generally referred to as simply a transducer or probe (the term transducer will be used throughout this commentary). The arrangement and the operating frequency of the crystal elements, as well as the width of the field of view (in metric) produced, are all taken into consideration when describing a transducer.

The arrangement, or array, of the elements within a transducer can be linear or curved (also referred to as “curvilinear”). A linear transducer contains many small rectangular crystal elements mounted side by side across its face. By triggering the elements sequentially, a rectangular image is built up from many vertical, parallel scan lines with a width that approximates the length of the array.⁷³ The advantage of a linear array is its wide near field, which is appropriate for imaging small superficial structures (FIGURE 7). A curved transducer is similar except that the crystal elements are formed into a curve rather than a straight line, which results in a di-

verging (pie or sector shaped) image (FIGURE 8). The advantages of a curved array is its wide far field, coupled with a small “footprint,” which is suitable for imaging deep abdominal structures.

A typical ultrasound transducer produces a range of frequencies around a preferred (maximum efficiency) frequency that is referred to as the “operating frequency” or “resonance frequency.” The operating frequency of an ultrasound transducer is predetermined by the thickness of the crystal elements. It is commonplace that a transducer may have 2 distinct operating frequencies (eg, 3.5 and 5.0 MHz, or 7.5 and 10.0 MHz) and, indeed, some are multifrequency.

BRIGHTNESS MODE AND MOTION MODE USI

THERE ARE SEVERAL OPTIONS (modes) available to display the electrical signal representing the ultrasound echo that returns from the tissues. The most common modes of display employed in rehabilitative settings are “B” (brightness, brilliance) and “M” (motion, movement) modes (b-mode and m-mode, respectively).

B-Mode USI

B-mode displays the ultrasound echo as a cross-sectional grey-scale image and is the mode of display most typically associ-

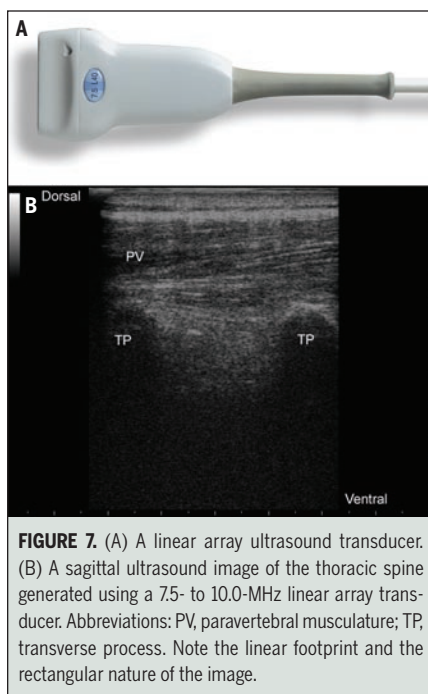


FIGURE 7. (A) A linear array ultrasound transducer. (B) A sagittal ultrasound image of the thoracic spine generated using a 7.5- to 10.0-MHz linear array transducer. Abbreviations: PV, paravertebral musculature; TP, transverse process. Note the linear footprint and the rectangular nature of the image.

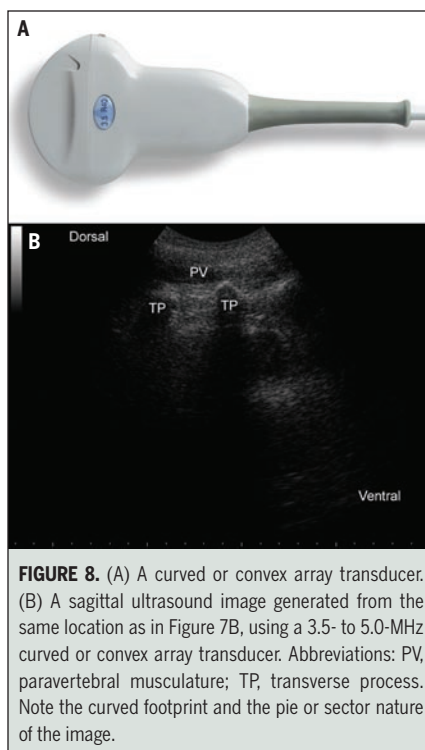


FIGURE 8. (A) A curved or convex array transducer. (B) A sagittal ultrasound image generated from the same location as in Figure 7B, using a 3.5- to 5.0-MHz curved or convex array transducer. Abbreviations: PV, paravertebral musculature; TP, transverse process. Note the curved footprint and the pie or sector nature of the image.

ated with USI (**FIGURE 9A**). B-mode images provide information gathered from the entire length of the transducer and consist of visible dots or pixels of varying degrees of brightness that represent the location and density of structures encountered by the ultrasound beam. The brightness of each pixel depends on the strength of the echo, which in turn is determined by the location and characteristics of the echo-generating structure. The position or plot of a pixel is established by considering the direction of an ultrasound wave when it enters the body, the length of time it takes for the echo to return to the transducer, and the speed at which sound can travel through soft tissue.¹³⁸

In contrast to other modes of display (eg, m-mode), the relatively large field of view available to b-mode, combined with the real-time nature of USI, presents an opportunity to view several structures at once and, if warranted, over time. Consequently, it can be used to depict the morphology (eg, shape, size, composition, and resting state) of a structure (eg, muscle, nerve), the positional relationship of several structures (eg, muscle, nerve, bone, or organs such as the bladder), as well as the characteris-

tics (simultaneous versus independent, or phasic versus sustained increase in muscle thickness) and the influence of a dynamic event, such as a muscle contraction, on structures within the field of view. Hence, it has been speculated that b-mode USI may be able to enhance clinical rehabilitative outcomes by contributing previously unavailable information about the structure and behavior of muscle to the examination process^{33,70,105,106,120} and by providing useful feedback about the behavior of muscle during therapeutic interventions.^{27,45,93} Furthermore, because of its advantages and capabilities, b-mode USI may have a role to play in basic, applied, and clinical rehabilitative research.^{9,20,52,57,88,124}

Clinical Applications of B-Mode USI: Evaluation Although used extensively in the laboratory, the clinical use of b-mode USI and the evidence base supporting it are in their infancy. That said, clinicians may look to related and emerging research to speculate on the kinds of information able to contribute to the examination process.¹⁴²

B-mode USI is well established as a

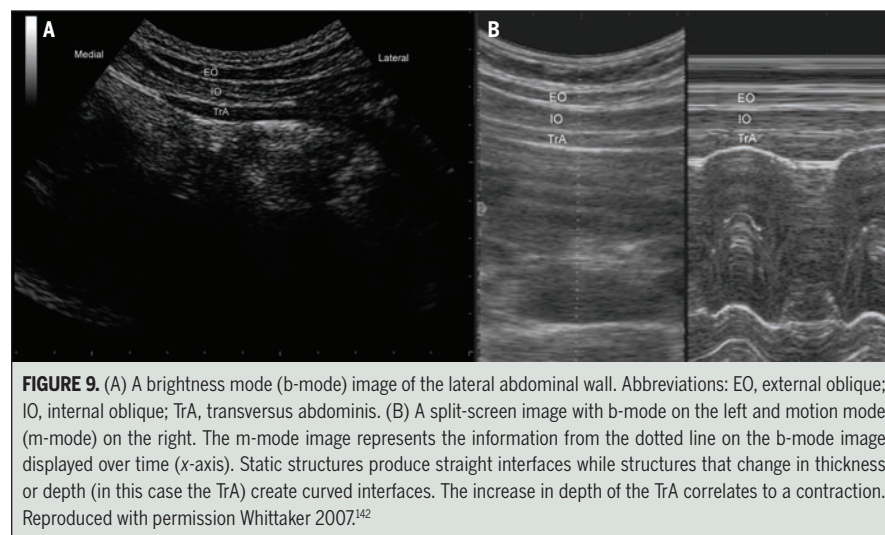
tool for measuring the static architectural features of a muscle, the positional relationships between muscles and/or other structures, as well as changes in these features and relationships over time. For instance, measurements of assorted muscle parameters (length, depth, cross-sectional area) for a wide variety of muscles, including the biceps brachii,⁶ masseter,¹⁰² cervical⁷⁷ and lumbar⁴⁸ multifidus, transversus abdominis,⁵² rectus abdominis,¹⁹ rectus femoris,⁶ supra and infraspinatus,⁶⁵ as well as the vastus lateralis,¹⁰⁷ have been validated through comparison to magnetic resonance imaging (MRI). Furthermore, USI has been used to describe the relationship between the pelvic floor muscles and the bladder wall,^{131,133} the bladder neck and symphysis pubis,¹⁰⁶ the bladder neck and anorectal angle,^{98,99} as well as the bladder base and the urethrovesical neck.²¹ As an extension of this work, investigators have been able to demonstrate acceptable interrater and intrarater reliability for various measurement applications,^{18,74,88,117,129} to generate normal reference ranges,^{104,105,120} to demonstrate differences in these parameters over time between normal and various patient cohorts^{51,100,130,133} as a result of therapeutic interventions,⁵⁰ and to investigate the relationship between the size and strength of specific muscles in varied populations.^{4,67,88,145,146}

Recently, the role that b-mode USI has played in detecting the presence of muscular degeneration resulting from aging and/or chronic dysfunction has been investigated. As muscular degeneration is associated with a decrease in water and an increase in fat and fibrous content,^{10,136} it results in greater echogenicity and a loss in the demarcation of a muscle's architectural features (muscle contour, pennate pattern, and the central tendon).^{66,124} Although MRI is considered the gold standard for identifying these changes, examples of these findings have been reported with b-mode USI for several muscles, including the cervical⁷⁴ and lumbar¹²⁰ multifidus, the rectus ab-

dominis,¹⁹ and the rotator cuff.¹²⁴ Furthermore, with the use of a qualitative evaluation tool that has incorporated both the degree of demarcation of architectural characteristics and muscle echogenicity with respect to a reference muscle at a set level of gain, Strobel et al¹²⁴ have concluded that b-mode USI is moderately accurate for the detection of significant levels of fatty atrophy in the supraspinatus and infraspinatus muscles when compared against MRI.

B-mode USI has also been used to comment on changes in architectural features of muscle, as well as the positional relationships of muscles or other structures in both normal and patient populations during dynamic events, such as a muscle contraction (voluntary and automatic) or increases in intra-abdominal pressure. Specifically, changes in architectural features of biceps brachii, tibialis anterior, transversus abdominis, and the internal and external oblique muscles have been investigated and compared to the amount of muscle activity present with electromyography (EMG) during voluntary contractions.^{57,82} Furthermore, automatic changes in these parameters have also been monitored during specific tasks. For instance, Ferreira et al³³ described changes in the depth and length of the lateral abdominal wall muscles during a lower extremity lifting task, while Kiesel et al⁷⁰ have described changes in the depth of the lumbar multifidus with a prone arm lift movement. B-mode display has also been used to monitor the position of the bladder base,¹³¹ the bladder wall,^{114,132,133} the bladder neck,^{20,106,133} and the anorectal angle^{98,99} during voluntary pelvic floor muscles contractions, increases in intra-abdominal pressure (Valsalva maneuver),¹³⁰ and lower extremity lifting tasks.⁹⁴

Although this work is valuable and has provided insight into mechanisms of neuromuscular dysfunction, it has also highlighted the limits to the information that USI can provide when considered in isolation. Specifically, as the relationship between actual muscle activity (measured



with indwelling EMG) and changes in the architectural features of a muscle (seen with USI) is nonlinear,^{53,57} a change in muscular dimensions may or may not indicate an increase in muscle activity. The relationship of these 2 factors is unique to each muscle; however, there is generally a rapid increase in muscle thickness associated with lower levels of muscle activity (approximately less than 25% of maximal voluntary contraction), which tapers as activity increases.⁵⁷ The discrepancy exists due to limitations of 2-dimensional imaging and the factors related to the length, pennation pattern, and extensibility of a muscle, as well as to the potential for a change in architecture in the presence of a competing force on the muscle (eg, contraction of an adjacent muscle or an increase in intra-abdominal pressure).⁵³ Similar issues arise when describing the relative change in position of a structure. Due to these considerations, investigators must take care during both the interpretation and reporting process. Ultimately, investigators must be adequately trained and experienced to be able to detect, reliably measure, and interpret the causes behind a change in one of these previously mentioned parameters. Furthermore, care must be taken to limit reporting to a change in the parameter measured (eg, thickness) and acknowledge that any conclusions regarding muscle activation or the mechanisms behind these

changes are an extrapolation of these findings. Nevertheless, if these issues are considered and appropriate care is taken, accurate analysis and measurement are possible.^{9,33,52,57,70,104,105,120}

Clinical Applications of B-Mode USI: Biofeedback The importance of coordinated muscle effort (neuromuscular control) has received considerable attention with respect to the rehabilitation of cervical and lumbopelvic dysfunction, as well as incontinence, in recent years. This is due to an accumulation of evidence pointing to altered neuromuscular control in individuals with persistent and recurring symptoms.^{5,15,24,56,60,62,126} Moreover, investigations indicate that these deficits do not consistently recover with the resolution of pain^{49,51} and are not addressed with traditional exercise programs focused on increasing strength and functional capacity.¹²⁵ The extrapolation of this work is that the initial focus of rehabilitation may need to address these motor control alterations through a therapeutic intervention rooted in motor learning.^{64,134,135}

As the real-time nature of b-mode USI allows a patient and therapist to view a muscle contraction and its impact on surrounding structures directly, it is a unique tool that may be a novel and previously unavailable resource to the learning process. First, it may serve as a tool that allows a therapist to explain and physically

demonstrate to a patient the subtleties of specific motor control impairment; second, it may serve as a comprehensive form of biofeedback providing knowledge of results and performance and enabling the modification of motor response. Although the literature is unclear as to how this knowledge might enhance motor learning or the permanence of these effects, recent findings^{27,45,93,139} suggest that real-time b-mode USI may enhance motor learning.

Research Applications of B-Mode USI Research applications of b-mode USI primarily involve assessment of the morphological characteristics of muscle (length, depth, diameter, cross-sectional area, volume) and changes in these characteristics, and the corresponding effect on associated structures (fascia and organs such as the bladder) with contraction, dysfunction, or therapeutic interventions, in an attempt to provide insight into the mechanisms that underlie alterations in the neuromuscular system. Paramount to the investigative process is the understanding that various factors influence the robustness and reporting of these measurements.

At the most fundamental level, this involves the need for standardization of imaging and measurement procedures. This includes definition of measurement site, definition of muscle borders, as well as matters related to repeated measurements, such as consistent patient positioning (eg, joint angle alters muscle cross-sectional area and length), transducer location, orientation, and inward pressure. Ideally, a repeatable transducer location is achieved through the use of bony or fascial landmarks that serve as standard reference points from which measurements can be taken at different points in time. If such reference points are not available within the ultrasound image, then carefully defined surface transducer locations generic between subjects^{33,105,129} or regions of the greatest visualized displacement of a structure (eg, the region of the bladder wall that exhibits the greatest displacement

during a pelvic floor muscles contraction)^{114,133} may be used. Relocation of the ultrasound transducer can also be aided by external markers (eg, freckles or scars) that can be traced onto a transparent sheet to form a map of the site and stored for future use.¹²² Furthermore, the sonographic convention in terms of positioning the ultrasound device on the right side of a supine subject (or left side of a prone subject) is recommended during research applications to aid in standardizing the orientation of the resulting images.¹⁶ However, this protocol may not always be feasible when assessing dynamic functional activities.

To facilitate comparisons between studies and the development of reference data for clinical purposes, it is suggested that future reports related to muscles size and other characteristics include mean, standard deviation, range, and 95% confidence intervals. Moreover, as these values have been found to vary based on gender and body mass index,^{105,117,120} comparison between individuals may be enhanced by standardizing the values across subjects by normalizing the postevent by the pre-event measurement and expressing this as a percentage.¹²⁹

With respect to statistical analysis, different tests have been used to investigate the reliability of USI measures, most commonly intraclass correlation coefficients (ICC)¹¹⁵ with standard error of measurement (SEM) and minimal detectable change (MDC).²⁸ Bland and Altman tests also provide a clinically meaningful measure of the magnitude of agreement (95% limits of agreement) independent of the true variability in the observations.⁷ These tests have their individual strengths and weaknesses, and no single test is sufficient to reflect reliability fully.^{101,103,115} It is recommended that future studies use all of these methods of analysis to enable comparison between reliability studies.

Although b-mode USI has limitations, it nonetheless appears to provide an opportunity to gather novel information. Hence, future work should focus on

determining its clinical utility as both an evaluative and therapeutic tool. Further, investigation should be undertaken into its ability to predict symptomatology, appropriateness for intervention, as well as categorization of subjects into homogeneous cohorts for interventional studies.

M-Mode Ultrasound Imaging

Unlike b-mode, which generates a cross-sectional image of an anatomical region using information gathered from the entire length of the transducer (**FIGURE 9A**), m-mode displays information collected from the midpoint of the transducer as a continuous image over time (as represented as the dotted line on **FIGURE 9B**). With time on the *x*-axis, and the depth of the underlying anatomical structure on the *y*-axis, the m-mode image represents changes in thickness, or depth of a structure, over time and is, therefore, referred to as “time-motion” mode. For example, the image in **FIGURE 9B** displays the change in thickness of the transversus abdominis muscle from a resting to contracted state over time.

Investigators have found m-mode USI to be a reliable technique to measure muscle thickness.^{9,69,82} Further, changes in muscle thickness measured by m-mode have been correlated to those generated by b-mode.^{82,137} McMeeken et al⁸² found acceptable agreement between the 2 display modes for measuring changes in thickness of the lateral abdominal wall muscles. The intrarater ICC value for b-mode was 0.989, for m-mode 0.981, and the between-mode reliability was 0.817.

M-mode also provides an opportunity to assess the depth of a structure over time and allows for the calculation of the relative timing of muscular thickness changes. For example, Mittal et al⁸⁶ used m-mode USI to assess the temporal relationships between circular and longitudinal muscle contractions during esophageal peristalsis. More applicable to physical therapy, Vasseljen et al¹⁴⁰ used high-frame-rate m-mode USI to detect the onset of lumbar multifidus

activity associated with limb motion, while Bunce et al^{8,9} found that m-mode USI was able to assess functional components of the lateral abdominal muscles during treadmill walking.

Until recently, m-mode has been a mode of display used almost exclusively in echocardiography to assess the structure and motion of the myocardium and the heart valves.^{34,87,143} Specifically, it has been used to assess morphological and functional changes in the myocardium during isometric exercise¹² and endurance training,^{38,42,112} as well as comparison of these changes between different populations (athletes,⁹⁶ nonathletes,⁹⁶ and obese individuals¹¹⁶) and as a function of age.⁹⁷

In addition to the above applications, m-mode USI has been used to assess pulmonary function, specifically diaphragm excursion, motility, and paralysis.^{2,32,39,78,113} Furthermore, it has been found to be beneficial in assisting and guiding treatment in those with muscular dystrophy. Researchers assessing diaphragmatic motion using m-mode USI have found that gender, body mass index, waist circumference, and age influence the amount of excursion.⁶⁸ These findings highlight the need to further assess the effects of these variables on measurement of muscular function when using m-mode for RUSI.

Although the use of m-mode USI in the assessment of muscle behavior is relatively new,^{8,9,69,82,140} it appears to have the potential to provide unique information. Specifically, it may assist investigators in describing changes in the function of the lateral abdominal wall, posterior spinal, and pelvic floor muscles associated with dysfunction.

HIGH-FRAME-RATE USI

CONVENTIONAL M-MODE ULTRASOUND images are constructed from data updated approximately 25 to 50 times per second. Although these frame rates are capable of detecting deformation (thickness) and changes in

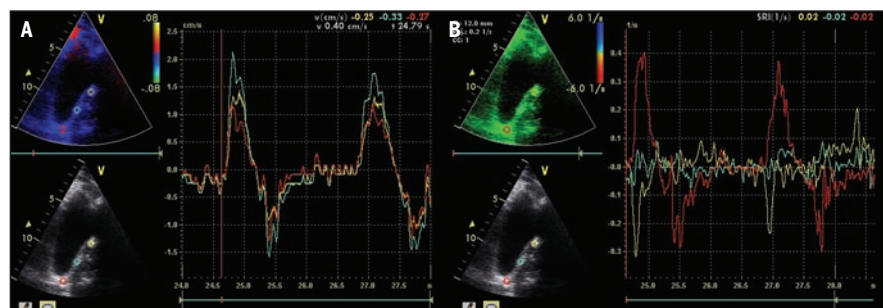


FIGURE 10. (A) Tissue velocity imaging. In the bottom left of the figure is a grayscale brightness mode ultrasound image of the bladder base during subsequent pelvic floor contractions. It is used to navigate and localize 3 measuring points (red, green, and yellow circles). Above this is the same image with tissue velocity analyzed via colour Doppler for the 3 sites. By plotting the velocity (cm/s) of the sites (x-axis) against time (y-axis), it can be determined that in this example there is no difference in velocity between the points. Scanner: Vivid 7, GE-Vingmed Ultrasound, Horten, Norway. Note the bladder is sparingly filled to better demonstrate displacement. (B) Strain rate imaging. Strain rate analyses (the rate by which strain occurs) of the same data involve plotting the strain rate period (1/s) against time (s). In this example strain rate differs at the 3 sites. The tissue marked by the yellow point has a negative strain rate, indicating that the tissue is compressed at the beginning of the contraction. The tissue marked by the red point initially exhibits a positive strain rate, indicating elongation; it then exhibits a negative strain rate indicating compression. The tissue marked by the green point undergoes mild compression and elongation (Scanner: Vivid 7, GE-Vingmed Ultrasound, Horten, Norway).

the depth of a muscle, they are not high enough to provide information related to the normal anticipatory response demonstrated by certain muscles^{23,54,60,89} and the loss of this response with dysfunction.^{24,55,58,79} In fact, to be able to record anticipatory muscle response (defined as a contraction occurring from 100 milliseconds before and up to 50 milliseconds after activation of a prime mover^{3,59,89}), frame rates need to be on the order of 500 frames per second.¹⁴⁰

Although intramuscular EMG is considered the gold standard for evaluating onset of muscle activity, high-frame-rate m-mode USI is a promising noninvasive alternative, as it allows for the visualization of the onset of deformation of muscle as it starts to contract. For instance, Vasseljen et al¹⁴⁰ demonstrated that high-frame-rate m-mode USI, captured at 500 frames per second, has comparable accuracy (when based on averaged values of repeated trials) to intramuscular EMG in detecting the onset of lumbar multifidus activity in healthy individuals. The superficial location of multifidus and the use of a high-frequency (12-MHz) transducer made the high frame rate and ultimately the investigation possible.

High-frame-rate m-mode USI, alongside methods such as tissue Doppler, falls into a category of imaging aimed at investigating tissue deformation, motion, and strain. As indicated above, high-frame-rate m-mode USI can be used to detect the onset of muscle deformation, as it shortens and thickens with a contraction. In contrast, tissue Doppler can be used to calculate tissue strain and strain rates,⁴⁴ as well as tissue velocity (FIGURE 10). Both deformation and velocity imaging can be derived from conventional ultrasound or tissue Doppler data; however, strain and strain rate require postprocessing.

Limitations of high-frame-rate USI vary across scanning devices. In general, the limitations of high-frame-rate m-mode are similar to b-mode in that they detect the earliest onset of motion induced by muscle contraction, whether as a result of actual contraction or the displacement of surrounding tissue. Furthermore, although contraction of a muscle produces displacement in 3 dimensions, m-mode applications are only capable of providing information about movement towards or away from the transducer.⁵³

Although the majority of studies employing high-frame-rate m-mode USI are

focused on describing the cyclic motions and deformation of the heart,¹²³ developments and experiences in the field of echocardiology hold potential for the description of a variety of parameters related to the contraction of skeletal muscles in rehabilitation. Specifically, research is required to determine whether high-frame-rate m-mode USI is helpful in investigating the location of the onset of a contraction within a muscle, differences in the onset of muscle activity within and between individuals or populations (symptomatic versus asymptomatic), and these at different points in time, as well as the sequence, timing, and patterns of muscle activation. These insights may provide valuable information to our understanding of automatic and voluntary muscle activity.¹⁴¹

ELASTOGRAPHY

IT IS POSSIBLE TO POSTPROCESS THE electrical signal produced from the echo returning from the tissues to the transducer in such a way as to quantify tissue movement and deformation in response to internal or external mechanical forces.^{13,92} In the last decade, several of these techniques (elasticity imaging and speckle tracking), including elastography, have been developed.

Elastography was initially conceived with a goal of quantifying the subjective information conveyed by palpation of harder areas within softer tissues, such as in the clinical detection of tumors (eg, breast, prostate).^{17,35} In the classical elastography method, the ultrasound transducer is used to slightly compress the tissue, while a rapid series of successive ultrasound images is acquired. Using cross-correlation methods to postprocess the ultrasound data, displacement and strain images are calculated and represent the amount of movement that small segments of individual A-lines (the electrical signal coming from a single transducer) undergo during the tissue compression.⁹² A-line segments with more relative motion correspond to softer tissue (assum-

ing a uniform strain distribution within the tissue).

The technique has since been extended to other applications, including the use of external sources of tissue motion (such as vibration), as well as naturally occurring internal motions (such as breathing, cardiac wall motion, and arterial pulsation).^{72,108} Furthermore, increased availability of high-frame-rate USI devices has allowed for the use of handheld transducers, which has simplified the frame-suspended setups used in earlier applications.⁷¹ Due to these advancements, clinical applications have expanded to include the detection of liver fibrosis and deep-vein thrombosis.^{96,109} Musculoskeletal applications include the quantification of soft tissue displacement and strain in response to a variety of externally applied mechanical inputs such as tension, compression, and acupuncture needle manipulation.^{37,41,76}

To date, tissue elasticity imaging techniques have not been used for rehabilitation purposes; however, they hold potential for the detection of differences in the biomechanical properties of muscle and its associated connective tissue in response to physical tasks. It is important to keep in mind that elastography images do not directly represent tissue elasticity but, rather, tissue displacement and strain. However, in conditions in which local tissue stress can be calculated (or estimated), strain and stress values can be used to map local tissue stiffness.

Although these elastic imaging techniques hold potential for rehabilitation, some practical difficulties need to be overcome. These include access to the raw electrical ultrasound signal (not available on most commercially available USI equipment) and the need for postprocessing of the ultrasound data, which preclude real-time feedback. Despite these limitations, the dynamic spatial mapping of tissue strain over time offers exciting new possibilities for quantifying the behavior of soft tissues in response to externally or internally generated perturbations.

USI VERSUS OTHER IMAGING METHODS

THE IDENTIFICATION, EVALUATION, and monitoring of various musculoskeletal disorders is expanding due to technological advancements associated with MRI, CT, and USI. Specifically, new and innovative applications are improving clinical understanding of the underlying mechanisms and sequelae common to musculoskeletal disorders.^{11,30,40,47,66,84} As such, it is important to consider how USI and the information that it can provide compares to these other imaging technologies.

MRI, CT, and USI provide insight into various features of the muscular system, both in asymptomatic as well as symptomatic individuals.^{31,49,70,77,105,120} In particular, they provide useful qualitative/quantitative measures concerning the muscular system, including the consequential muscular degeneration (atrophy and fatty infiltrate) shown to be common in patients with low back pain,^{11,22,47,51,66,75,84,95} neck pain,^{1,30,40,74,83} and other peripheral musculoskeletal disorders.¹²⁴ Although MRI is considered the gold standard for musculoskeletal imaging, emerging applications of USI and CT are capable of providing insight into *in vivo* features of the musculoskeletal system. Each imaging method has strengths as well as weaknesses (TABLE).

Magnetic Resonance Imaging

MRI, unlike USI, has multiplanar and multislice imaging capabilities. It is considered the gold standard for the evaluation and quantification of soft aqueous muscular degeneration, as it provides reliable measures of degenerative changes in muscle such as fatty infiltration and atrophy.^{1,11,30-31,40,52,66,83,84,90,95} There are 2 conventional MRI sequences: T1 and T2 weighted. Images from T1-weighted scans demonstrate excellent anatomical contrast of fat and other soft aqueous tissues (eg, skeletal muscle).⁹⁰ Alternatively, T2-weighted scans provide outstanding detail related to the features

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of inflammation that are suggestive of neopathological conditions. The drawbacks of MRI remain cost, accessibility, constraints in the number of joints that can be investigated per session, limited real-time imaging capacity, and variable patient tolerance (eg, claustrophobia, metallic implants, pacemaker, and pregnancy).

Computerized Tomography

CT, like MRI but unlike USI, permits multislice imaging and can offer better scan resolution and shorter imaging times than MRI. However, it is not without the inherent risks associated with exposing a patient to ionizing radiation. CT is useful in diagnosing traumatic musculoskeletal injuries, such as fractures, and has been effectively used to evaluate and quantify cross-sectional area of paraspinal musculature in patients with low back pain.^{22,75} While CT produces high-quality images, they are dependent on tissue densities in order to provide contrast. When tissue densities between pathologic and adjacent anatomy are similar, contrast media may be required for differentiation, rendering CT inadequate if a patient has a history of contrast reaction.⁸⁰

Ultrasound Imaging

USI, although less sophisticated in terms of resolution than MRI and CT, has advantages as a safe, cost-effective, portable, and clinically accessible method for gathering information about the static characteristics of muscle,^{47,51,70,74,105,120,124} as well as muscle behavior during dynamic events.^{50,77} As such, it shows promise as a tool in musculoskeletal examination and treatment. Moreover, unlike CT, USI does not expose the patient to ionizing radiation and is well tolerated by patients. A feature unique to USI is its dynamic capability of scanning in real time, which makes it superior to MRI and CT for evaluating mobile structures such as tendons, nerves, and muscles, and it may become an important tool for directing appropriate physical therapy treatment decisions.¹²⁷ However, as high-

COMPARISONS BETWEEN DIFFERENT IMAGING MODALITIES USED IN THE ASSESSMENT OF MUSCULOSKELETAL DISORDERS*

	MRI	USI	CT
Cost	Expensive	Inexpensive	Intermediate
Ease of accessibility	Difficult	Easy	Difficult
Ionizing radiation	None	None	Yes
Supports intervention	Yes	Yes	Yes
Operator-dependent	No	Yes	No
Imaging capability			
Planes	Multi	Variable axes to joint surface	Multi
Anatomy			
Muscle	Excellent	Good	Fair
Fat	Excellent	Fair	No role
Tendons and sheaths	Good	Excellent	No role
Ligaments	Good	Excellent	No role
Synovial membrane	Good	Excellent	No role
Bone	Excellent	Good	Good
Cartilage	Good	Excellent	Fair
Inflammation	Excellent	Good	No role
Number of joints/session	Few	Many	Few
Real-time scanning	Cardiac only	Yes	No
Patient tolerance	Variable	Good	Variable

*Adapted from Tan et al²⁷ with permission.

Abbreviations: CT, computerized tomography; MRI, magnetic resonance imaging; USI, ultrasound imaging.

lighted throughout this commentary, USI is not without disadvantages and is highly operator dependent. Perhaps the most promising feature of USI is its accessibility and the feasibility for physical therapists to acquire the skills needed to incorporate its use into clinical practice. However, evidence for its use in different applications within rehabilitation is needed before widespread routine clinical use can be promoted.

FUTURE CONSIDERATIONS

IT IS UNCLEAR WHETHER THE EVALUATION of muscular and motor control abnormalities differs across imaging technologies, muscle, body regions, body side, varying diagnoses, and assorted anthropometric variables. Further, the value of USI from a pathoanatomical and pathophysiological perspective, although under investigation, has yet to be determined. Although innovations in muscular imaging research are enhancing our

understanding of muscle dysfunction, degeneration, and control, and slowly influencing clinical practice, there is a need to standardize techniques that are cost effective, reliable, easily accessible, and well tolerated by both patients and clinicians in order to ensure their appropriate use.

Future research efforts should address the diagnostic and prognostic validity of USI in patients with acute musculoskeletal pain and a wide variety of musculoskeletal disorders by comparing USI, MRI, EMG, and perhaps CT results. Randomized control trials are also needed that include comparison of interventions incorporating RUSI to those that do not to examine whether USI biofeedback improves outcomes. Ultimately, such studies could provide appropriate evidence-based evaluation and treatment strategies that incorporate RUSI as an outcomes measure. Routine adoption of RUSI in physical therapy requires appropriate training programs.

CONCLUSION

THE GOAL OF THIS COMMENTARY HAS been to provide an overview of basic USI and instrumentation principles, including an understanding of the various modes and applications of the technology with respect to neuromusculoskeletal rehabilitation and in relation to other common imaging modalities. In doing so, we hope that the reader has gained a greater understanding of the value of this tool both from a clinical as well as an investigative perspective. Although other imaging modalities may be superior in providing some, although not all, of the information available with USI, there is growing access to, and evidence in support of, its use by physical therapists. As such, it is imperative that there is support for further investigation with regard to its potential and also a greater understanding of its limitations, as it is likely that the full significance of USI in relation to rehabilitation has yet to be established.

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APPENDIX

GLOSSARY OF TERMS

Acoustic shadowing—The reduction of sound wave echoes from structures that lie behind a strongly reflecting or attenuating structure (eg, bone)⁷³

A-line—An electrical signal corresponding to the scattering and reflection of ultrasound from tissue or other media, generated from a single ultrasound transducer either used alone, or as part of a linear (brightness-mode) array

A-line segment—A small portion of an A-line used in elastography cross-correlation analysis

Artifact—Incorrect representations of anatomy or motion (eg, situations that result in structures that are not real, missing, improperly located, or of inaccurate brightness, shape, or size). Examples include acoustic shadowing, edge shadowing, enhancement, and reverberation¹⁴²

Attenuation—The reduction in the intensity or amplitude of a sound wave, expressed in decibels (dB/cm¹/MHz¹) and caused by absorption, scattering, and reflection of the sound wave as it travels. As attenuation increases, penetration decreases⁷³

B-mode—Brightness or brilliance mode

Cavitation—Refers to the production and behavior of gas bubbles within a liquid when exposed to the sound wave. This behavior can be variable (eg, oscillation or collapse) and depends upon factors such as the size of the cavity, and the nature of the immediate environment⁹¹

Doppler imaging—The basic principle of Doppler ultrasound lies in the observation that the frequency of a sound beam reflected back to its source is altered when it encounters a moving object.¹³⁸ As the change in frequency is proportional to the velocity of the object, Doppler imaging can be used to display flow (blood) or tissue velocity

related information

Echogenic—A structure or material that produces echoes (eg, reflection of ultrasound waves). The more echogenic a structure or substance is, the whiter it will appear within an ultrasound image¹⁴²

Edge (refractile) shadowing—Refers to specific type of acoustic shadow that is generated when a sound wave encounters an object with a curved surface (eg, bladder or cyst). The shadow is observed at the lateral margins of the object where the sound beam contacts the interface at a very oblique angle. As a result of both refraction and reflection, none of the incident sound returns to the transducer from this region and a shadow results

Enhancement—The strengthening of a sound wave echo distal to a weakly attenuating structure (eg, a fluid-filled organ such as the bladder)⁷³

Far-field—The bottom half of the ultrasound screen, which represents that part of the body furthest from the ultrasound transducer¹⁴²

Field of view (FOV)—Refers to what is visible on the ultrasound display FOV and is dictated by the shape (curvilinear versus linear) and width of the transducer, as well as the depth setting of the image display. The ultimate depth of the FOV is determined by the frequency of the transducer, the power setting of the ultrasound device, as well as the characteristics of the medium that is being imaged

Frame rate—The number of frames of echo information stored each second⁷³

Frequency—The number of oscillations a molecule or a sound wave undergoes in 1 second. Frequency is expressed in Hz¹⁴²: 1 Hz = 1 cycle per second; 1 kHz = 1000 cycles per second; 1 mHz = 1000 000 cycles per second

Gain—Refers to amplification (expressed in dB) of the echoes returning from the tissues back to the transducer. The degree of amplification is under the voluntary control of the operator.^{73,142}

Hyperechoic—A structure or substance that is more echogenic (whiter and brighter on the ultrasound screen) than surrounding tissue. The surface of bone and dense fascia are examples of hyperechoic media.¹⁴²

Hypoechoic—A structure or substance that is less echogenic (darker on the ultrasound screen) than surrounding tissue. Fluids such as blood and urine are examples of hypoechoic media.¹⁴²

Impedance—The resistance that a tissue or medium has to sound. Acoustic impedance depends upon the density of the medium and the speed at which sound can travel through it. It is expressed in rays.⁷³

Incidence angle—The angle between the sound coming from the transducer (incident sound) and a line perpendicular to the boundary of a medium.⁷³

Intensity—The rate at which energy is delivered per unit area.⁷³ The intensity of an ultrasound wave is determined by the total power output of the transducer (W) divided by its effective radiating area (cm^2) and expressed in units of mW/cm^2 .

M-mode—Motion mode, sometimes referred to as time-motion (TM) mode.

Morphology—The study of form and structure.²⁵ With respect to muscle, this refers to describing characteristics of its dimensions (eg, cross-sectional area, length, shape ratios, and depth/thickness), as well as tissue composition.

Morphometry—The quantitative measurement of form.¹¹¹ With respect to muscle, this refers primarily to measurement of its dimensions (eg, cross-sectional area, length, shape ratios, and depth/thickness). Morphometry does not consider tissue composition.

Motor control—“An area of study dealing with the understanding of the neural, physical, and behavioral aspects of movement.”¹¹¹ Relates to the timing, magnitude, and sequencing of muscle activation and relaxation.

Motor learning—A set of internal processes associated with practice or experience leading to relatively permanent changes in the capability for motor skill.¹¹¹

Muscle behavior—Observable activity or the response of muscle during a specific event or given set of circumstances. The behavior of muscle can be described by changes in its electrical properties over time or architectural characteristics (cross-sectional area, length, depth/thickness, or relationship to adjacent structures).

Muscle cross-sectional area (CSA)—A quantitative 2-dimensional measure of the plane of a structure created by cutting through it transversely.²⁵ Anatomical CSA refers to the CSA at 90° to the long axis (or direction) of the muscle fibers. Physiological CSA is not measured 90° from the long axis of the muscle fibers. Physiological CSA is commonly generated with rehabilitative ultrasound imaging (RUSI), for instance with muscles that have complex internal architecture such as the lumbar multifidus in which the fascicles pass in a caudal-lateral direction over the vertebrae.

Muscle inhibition—Reflex inhibition is the reduction or elimination of muscle activity associated with afferent stimuli from joint receptors that reduce activation of alpha motor neurons in the anterior horn of the spinal cord (eg, reflex inhibition of quadriceps has been demonstrated in the presence of knee joint damage in the absence of pain).¹²¹ Voluntary inhibition refers to an unwillingness to contract a

muscle due to pain or fear of pain.

Muscle thickness (depth)—A quantitative linear measure from the superficial to deep aspects of a muscle.²⁶ Baseline or resting thickness refers to the thickness of a muscle in a perceived resting state.

Near-field—The top half of the ultrasound screen, which represents that part of the body closest to the ultrasound transducer.¹⁴²

Operating frequency—The preferred (maximum efficiency) frequency of operation of a transducer.⁷³ The operating frequency can also be referred to as the resonance or main frequency.

Penetration—Refers to the ability for sound to travel through media, thereby influencing image depth. Penetration is dependent upon the strength (intensity) and frequency of the sound wave, as well as the compressibility of the medium that it travels through. In descriptive purposes, penetration refers to image depth.^{73,142}

Pixel—An abbreviation of “picture element.” A pixel refers to the smallest unit of a digitized, 2-dimensional image. A pixel can be described by its location (a set of x and y coordinates), as well as its brightness.⁷³

Piezoelectric effect—A phenomenon in which some materials (ceramic, quartz, etc) produce a voltage or electrical current when deformed by an applied pressure, such as sound.⁷³

Real-time ultrasound imaging (RTUS)—The rapid sequential display of ultrasound images resulting in a moving presentation.^{73,142}

Reflection—As a sound wave propagates it attenuates (loses energy). Reflection is one form of attenuation and refers to the portion of the sound wave that is reflected back towards the source of the sound. This reflected wave is received and processed to generate an ultrasound image.^{73,142}

Refraction—Refers to the change in direction of a wave when it crosses a boundary. It comes from the modification of a Latin term meaning “to turn aside.”⁷³

Resolution—A measure of the ability of an instrument to show detail.¹³⁸

Scatter—Describes the generation of secondary waves (fractions) in response to the primary sound wave encountering a rough surface or heterogeneous media. Scattering is often referred to as diffusion.^{73,142}

Sonography—The term used to describe imaging resulting from ultrasound. The Latin word *sonus* is sound, and the Greek word *graphien* is to write.⁷³

Sound—Mechanical energy that propagates through air, water, or any other matter in an orderly, rhythmic fashion, as determined by the molecular makeup of the transmitting medium.^{73,138}

Strain—Mechanical deformation of a structure as the result of stress.

Strain rate—Strain rate refers to the instantaneous strain (or change in strain) per time unit. The strain rate has the same direction as the strain (eg, negative strain rate during shortening, positive strain rate during elongation).

Transducer—Any device that converts one form of energy into another. The piezoelectric crystal is a transducer that converts electrical energy into sound energy and vice versa.⁷³

Transducer assembly (commonly referred to as a transducer)—Consists of the transducer elements, their associated casing, and dampening material.^{73,138}