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Magnetic Resonance Elastography

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Abstract

Often compared to the practice of manual palpation, magnetic resonance elastography is an emerging technology for quantitatively assessing the mechanical properties of tissue as a basis for characterizing disease. The potential of MRE as a diagnostic tool is rooted in the fact that normal and diseased tissues often differ significantly in terms of their intrinsic mechanical properties. MRE uses magnetic resonance imaging (MRI) in conjunction with the application of mechanical shear waves to probe tissue mechanics. This process can be broken down into three essential steps:

1. inducing shear waves in the tissue,
2. imaging the propagating shear waves with MRI, and
3. analyzing the wave data to generate quantitative images of tissue stiffness

MRE has emerged as a safe, reliable and noninvasive method for staging hepatic liver fibrosis, and is now used in some locations as an alternative to biopsy. MRE is also being used in the ongoing investigations of numerous other organs and tissues, including, for example, the spleen, kidney, pancreas, brain, heart, breast, skeletal muscle, prostate, vasculature, lung, spinal cord, eye, bone, and cartilage. In the article that follows, some fundamental techniques and applications of MRE are summarized.

Keywords

elastography; elasticity; mechanical properties; shear stiffness

Introduction

The simple, yet enduring practice of palpation remains an important diagnostic tool for physicians, providing practitioners with a means to evaluate the mechanical properties of tissues and to distinguish between normal and abnormal states. The mechanical properties among specific tissue types are known to vary considerably and to be altered significantly under various physiological and pathological conditions, thus providing extensive opportunities for diagnostic evaluation [1,2]. In particular, palpation has long formed the basis for tumor detection in cancers of the breast [3], prostate, and thyroid, and in the detection of certain diffuse changes in tissue stiffness, such as cirrhosis of the liver. Despite

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its universal application, however, palpation is ultimately a subjective and qualitative practice, practical for only superficial tissues, since many tissues of the body, such as the brain, cannot be directly palpated non-invasively.

The advent of medical imaging has revolutionized our ability to acquire diagnostic information non-invasively, however, none of the fundamental contrast mechanisms provided by conventional imaging modalities, such as computed tomography (CT), ultrasound (US), or magnetic resonance imaging (MRI), are inherently sensitive to the properties evaluated by palpation. This has motivated the development of imaging techniques capable of providing quantitative information about the mechanical properties of tissues.

In clinical medicine, a variety of terms are used to describe the mechanical properties of tissues, such as “hardness” and “compressibility”. From an engineering standpoint, these qualitative properties can be described with quantitative “moduli” that relate stress (force per unit area) and strain (a unitless measure of deformation). More specifically, measures of the shear modulus (the proportionality constant in the transverse stress-strain relationship) and shear wave speed have been determined by many researchers to be the most useful for imaging the mechanical properties of tissues and for characterizing pathology. For more detailed information on tissue elasticity theory, the reader is referred to the following comprehensive guides [1,4].

Over the past twenty years, numerous techniques for imaging tissue elasticity have been developed and investigated, characterized primarily by the diversity of methods employed to a) excite the tissue mechanically, b) measure the tissue response, and c) analyze the measured response. Excitation methods include static [5–8] and dynamic approaches [9–13]; measurement techniques include US [9,14–18], MRI [6,11,19–22] and optical methods [23–28]; and types of analysis include qualitative methods [5,29,30] and a wide range of quantitative methods [11,31] derived from various rheological models. Today, the most common MRI-based technology for imaging the mechanical properties of tissue is known as magnetic resonance elastography (MRE), the subject of this review article.

Magnetic Resonance Elastography

Magnetic resonance elastography (MRE) is a dynamic, MRI-based technique capable of acquiring and converting images of propagating shear waves into maps of shear modulus (or stiffness). To date, this technology has been applied to a variety of tissues and organs, but has been most successfully used in the clinical detection of hepatic fibrosis, a common end-stage for a variety of liver conditions corresponding to a marked increase in liver stiffness [32,33]. Recently, MRE was awarded FDA clearance, and it is now available as an added feature on some new and existing whole-body MRI scanners.

MRE involves three essential steps: First, shear waves in the audible frequency range (10 – 1000 Hz) are introduced into the tissue with an external mechanical driver. Second, the propagating shear waves are imaged using a suitable MRI pulse sequence. Third, the resulting wave images are analyzed in order to yield quantitative measures of the tissue

stiffness. The sections that follow are an attempt to elaborate on these steps and to provide an introduction to some applications of MRE that are currently under development.

Mechanical Excitation

Typically, an external source of motion, called a driver, is used to introduce mechanical shear waves into the tissue of interest. In most cases, MRE is performed with a single frequency of motion that lies within the audible frequency range (10 – 1000 Hz). This sinusoidal waveform is digitally synthesized and synchronized with the imaging pulse sequence. Before being applied to the driver, the waveform is amplified to a power level on the order of 10 to 100 W, depending on the specific application and the efficiency of the driver used.

Numerous driving mechanisms have been devised for MRE, each well-suited for particular uses, and each with its own limitations [34]. One of the oldest designs is an electromagnetic actuator [11,35] that generates vibrations with a small voice coil placed within the polarizing magnetic field of the MRI scanner. Although simple in design and flexible in its ability to produce a pure shearing motion, these devices typically provide limited power and can introduce imaging artifacts during sensitive imaging sequences, such as echo planar imaging (EPI). Various configurations of piezoelectric elements have also been used to mechanically excite tissue, including piezoelectric benders, extenders and stacks [36–38]. Like electromagnetic actuators, piezoelectric devices can also produce well-controlled motion, but they tend to be fragile, low-power (high-voltage) and not widely used due to low relative displacements and high acoustic impedance. Piezoelectrics have also been used in the form of focused ultrasound (FUS) transducers (operating in the MHz frequency range) to generate radiation force at a small focal point, causing shear waves to propagate within the tissue [39–41].

Another more recent, but widely used approach for exciting motion employs a high-powered actuator located outside the magnet bore, typically in the form of a conventional acoustic speaker-like system, where the combination of a permanent magnet and voice coil is used to generate vibrations that can be conducted to the patient through a long rod [42] or air-filled tube [33]. In the latter case, commonly referred to as a pressure-activated driver, acoustic waves are driven into an enclosed air cavity, conducted to the patient through a long, flexible tube, and coupled to the patient with a passive, drum-like driver (Figure 1). Locating the active component of the mechanical driver remotely is a beneficial approach because it minimizes potential electromagnetic interference (resulting in image artifacts) by locating electromagnetic components outside of the magnet bore away from the region of interest. As indicated, the remote location also means that these devices can be scaled to deliver more power, making them suitable for vibrating larger anatomy of the chest and abdomen. The pressure-activated driver is easily placed during the patient examination and can be readily adapted for different applications, such as the breast, brain or extremities, by modifying the passive component. The pressure-activated driver is now available commercially and is currently used by a number of groups for clinical liver MRE applications [32]. The vibration amplitudes used in MRE are on the order of hundreds of microns, and have been shown to

meet regulatory standards for occupational exposure to whole-body and extremity vibrations [43].

Motion Encoding

The underlying MR imaging fundamentals of MRE are beyond the scope of this work, but for more information the reader is referred to the following comprehensive references [44–46]. Dynamic MRE utilizes a highly sensitive phase-contrast-based imaging method to image microscopic tissue displacements introduced into the tissue by one of the mechanical drivers mentioned in the previous section. This imaging approach was originally developed by Muthupillai et al [11], and uses bipolar motion-encoding gradients (MEGs) to encode the cyclic tissue displacement into the phase of the MR image, a technique that is closely related to the phase-contrast method originally developed for imaging blood flow [47]. As the induced mechanical wave propagates through the tissue, the physical displacement is encoded into the phase ϕ of the resulting MR image according to the following relation, given as a function of spatial position $\vec{r}$ and “phase offset” θ , a measure of the relative phase between the imaging sequence and the sinusoidal mechanical excitation:

$$\varphi(\vec{r}, \theta) = \frac{\gamma NT(\vec{G} \cdot \vec{\xi}_0)}{2} \cos(\vec{k} \cdot \vec{r} + \theta) \quad [1]$$

On the right side of the equation, γ refers to the proton’s gyromagnetic ratio ($\gamma/2\pi = 42.57$ MHz/T), N is the number of MEG pairs, T is the period of the MEG, $\vec{G}$ is its amplitude and direction, $\vec{\xi}_0$ is the amplitude and direction of motion, and $\vec{k}$ is the acoustic wavenumber (usually equal to 1). The relative phase θ between the applied motion and the MEGs is typically varied in equal steps of at least 4 phase offsets over the course of a complete cycle of motion. In most cases of MRE, motion is applied continuously, but when transient phenomenon are of interest, an additional time delay can be introduced to track the propagating wave beyond a single period of motion. In instances of both transient and harmonic (steady-state) MRE, recording multiple offsets permits easy visualization of the propagating wave through time, and allows post-processing in the temporal dimension. As equation 1 indicates, MRE can be made to be very sensitive, capable of encoding displacements on the order of hundreds of nanometers, subject to specific pulse sequence parameters and gradient hardware performance [11].

The primary distinguishing feature of an MRE imaging sequence is the presence of the MEGs. As such, MRE can be performed with a modified version of any number of existing MR imaging techniques. Shown in Figure 2 is a conventional two-dimensional gradient-recalled echo- (GRE) based MRE sequence, including radiofrequency (RF), slice-select (SS), phase-encode (PE), readout (RO), and motion waveforms. In practice, motion encoding is always performed between RF excitation and the RO interval, and any component of motion can be imaged by manipulating the direction of the MEG vector. In the simple example of Figure 2, the MEG is applied along the SS axis, which will encode motion in the through-plane direction. Like other phase-contrast MR imaging methods, two images are typically acquired (per phase offset) with opposing MEG polarities in order to

calculate a phase-difference image, effectively eliminating background phase information due to non-motion-related sources. The resulting images are also commonly referred to as wave images, and form the basis for the analysis described in the following section.

In addition to GRE-based MRE, a number of other conventional MR imaging sequences, including spin echo (SE), echo planar imaging (EPI), and balanced steady-state free precession (bSSFP), have been adapted for MRE [48–51]. Usually, these sequences employ MEGs with periods that match the applied mechanical excitation, and are therefore optimally sensitive to a single frequency [11]. MEGs can be designed, however, to encode for multiple frequencies [42,52], or to allow for other imaging considerations, such as shorter echo times (more signal) for short T2-species [53], in exchange for a relative decrease in motion sensitivity.

Mechanical Property Estimation

Once a series of wave images has been acquired using a suitable MRE sequence, an appropriate method of analysis must be selected in order to estimate the underlying material properties of the tissue (such as shear stiffness). For a material in the general case (homogeneous, viscoelastic and anisotropic), the equation of motion relating the applied stress to resultant strain is quite complex, and is expressed as a rank 4 tensor that includes 21 independent complex quantities [54]. Making the assumptions of isotropy, linear elasticity, homogeneity and planar wave propagation, however, simplifies the stress-strain relationship considerably, yielding the following solution for the “effective” shear modulus μ commonly used in dynamic MRE:

$$\mu = \rho c_s^2 = \rho (\lambda f)^2 \quad [2]$$

where ρ is the density (typically assumed to be 1000 kg/m³), c_s is the shear wave speed, λ is the wavelength, and f is the operating frequency [55]. Because the operating frequency is known, and the tissue is assumed to have the same density as water, the task of solving for the effective shear stiffness is reduced to measuring the local spatial wavelength. This process can be as simple as a manual estimate of wavelength using a one-dimensional profile of the wave field, but as the field of MRE has progressed, a number of algorithms have been developed for automatically measuring the wavelength, such as local frequency estimation (LFE) and the phase gradient (PG) method [56]. Although the stiffness measure reported by Equation 2 has been shown to be clinically useful, a variety of more sophisticated mathematical techniques, such as direct inversion and finite element analysis, have been employed to solve the wave equation directly using fewer simplifying assumptions [55,57,58]. Some of these techniques are capable of solving for the true complex shear modulus (where the imaginary component represents viscosity), and may provide additional independent parameters for characterizing tissue.

Images of the effective shear stiffness produced by the algorithms listed above are commonly called elastograms. In the sections that follow, the elastograms depict effective shear stiffness at a single operating frequency. Although MRE is typically performed at normal imaging resolutions, with voxel dimensions on the order of 1 mm, the resolution of

the typical elastogram is somewhat reduced (by a factor of 3 to 5) depending on the local spatial wavelength and specific processing employed. In addition, various preprocessing methods may be applied to the raw wave image data prior to inversion to limit artifacts in the elastograms. Phase unwrapping is commonly used to correct high-amplitude regions of the wave image, where large displacements have produced phase accumulations of more than $\pm\pi$ radians. Lowpass filtering may be used to limit the presence of noise, and highpass filtering or curl filtering may be used to minimize the presence of (low spatial frequency) longitudinal waves and bulk tissue motion. In addition, a technique known as directional filtering is sometimes used to minimize the effects that destructive wave interference can have on the inversion process [59].

Applications of MRE

The field of MRE is developing rapidly due to the versatility of the technology and its great clinical potential as a noninvasive diagnostic tool. Numerous possible applications have emerged as MRE has been used to investigate a variety of organs and tissues including liver, spleen, kidney, pancreas, brain, heart, breast, skeletal muscle, prostate, vasculature, lung, spinal cord, eye, bone, cartilage, heel fat pads and others [33,37,42,60–74]. Figure 3 contains a graphic summary of some of the effective shear stiffness results reported for MRE, including tissue types and operating frequencies, illustrating the range of stiffness values (orders of magnitude) available to this technique and its potential as a sensitive diagnostic tool. In the sections that follow, several current applications of MRE are presented and discussed.

Like most medical imaging experiments, phantom testing is often used in MRE to test motion-sensitive imaging sequences and wave image inversion methods. Data from an exemplary phantom experiment is displayed in Figure 4, showing a tissue-mimicking phantom comprised of regions of differing stiffness. In this example, the background material is made of 10% bovine gelatin, and the four cylindrical plugs are made of stiffer 1.5% agarose gelatin, easily visible as the hypointense regions in the magnitude image (Figure 4a). Harmonic shear waves are induced in the phantom at a frequency of 90 Hz using an electromagnetic actuator, illustrated schematically in Figure 4a, its direction of motion indicated by the two-sided arrow. The resulting wave image (Figure 4b), acquired with an MRE pulse sequence, displays tissue displacements in the left-right direction. Wavelengths in the stiff agar inclusions are observed to be longer than in the relatively soft background material. With an appropriate inversion algorithm, the wave data is converted into a shear stiffness elastogram (Figure 4c), given in units of kPa. Obvious contrast between the background material and the stiff inclusions is observed. Incidentally, also observed in the elastogram, is a difference in the stiffness of the two-part background material (which was which was poured in stages), a difference that is invisible in the magnitude image. Quantitative stiffness measurements for each region can be measured by placing a region of interest and calculating the mean value of the enclosed pixels.

Liver

By far the most widely adopted application of MRE to date has been the clinical identification of hepatic disease. Chronic liver disease is a growing global health problem, and liver damage due to hepatic fibrosis is a progressive process common to a wide variety of liver injuries. If left untreated, hepatic fibrosis can eventually lead to irreversible cirrhosis of the liver with fatal consequences. The current gold standard for diagnosing hepatic fibrosis is the needle liver biopsy, an expensive, invasive technique, associated with complications, and severely limited by sampling error [XX refs]. MRE of the liver provides a promising potential alternative to needle biopsy, because it is capable of detecting elevated liver stiffness, which has been shown to correlate strongly with the presence of fibrosis [32].

At our institution, clinical liver MRE is currently performed at 60 Hz using a pneumatic pressure-activated driver, four phase offsets, and a multi-scale direct-inversion algorithm for estimating liver stiffness. In Figure 5, clinical MRE examples are presented from a normal patient (top row) and a second patient with alcoholic cirrhosis (bottom row). Magnitude images of the normal and diseased livers are seen in Figures 5a and 5d, respectively, and outlined with dotted lines. These images yield no evidence for the presence or absence of liver disease in either patient. In the wave images (Figures 5b and 5e), however, the shear wavelength in the cirrhotic liver is observed to be significantly longer than that of the normal patient, indicating that the cirrhotic liver is considerably stiffer than normal. The corresponding elastograms (Figures 5c and 5f) confirm this qualitative finding. The mean stiffness values of the normal and abnormal livers were measured as 2.12 and 13.47 kPa, respectively.

Liver stiffness has been shown to correlate well with fibrosis stage, as seen in Figure 5g, where normal subjects and fibrosis patients are well-distinguished, and liver stiffness is seen to increase with disease progression. ROC analysis of the patient population presented in Figure 5g, yields an optimal diagnostic threshold of 2.93 kPa for distinguishing healthy and fibrotic livers, with overall accuracy of 99%, and sensitivity and specificity of 98% and 99%, respectively [33]. Unlike conventional imaging modalities, such as CT, US and conventional MRI, MRE is capable of detecting the otherwise “invisible” presence of liver fibrosis at a very early stage, long before the disease has progressed to irreversible cirrhosis [75,76]. The growing amount clinical evidence in this area suggests that MRE has great potential to provide a safe, comfortable alternative to needle biopsy that is also more accurate and less expensive.

The characterization of hepatic tumors is also an ongoing area of interest in the field of MRE. Malignant tumors have been shown to be measurably stiffer than benign tumors and normal liver parenchyma, and 5 kPa has been found to be a useful diagnostic threshold for differentiating between malignant and benign tumors and normal liver tissue [77]. In addition to the liver, preliminary work has been performed and continues on other organs of the abdomen, such as the spleen, kidneys and pancreas [XX refs].

Breast

Routine manual palpation of the breast has long been considered an important component of breast cancer screening [3] because malignant breast tumors are known to be stiffer on average than benign lesions and healthy breast tissue [78]. In recognition of the value of palpation, and owing in part to limitations with mammography, MRE has been applied to the problem of breast cancer diagnosis [72,79]. Specifically, MRE is being investigated as a complement to contrast-enhanced MR imaging (CE-MRI), which has very high sensitivity for the detection of tumor nodules, but unacceptable specificity, resulting in false positives and unnecessary biopsies [80]. It has been shown that MRE can provide additional information about lesions detected with CE-MRI, thereby improving the specificity of the combined technique [81].

In Figure 6, MRE data is presented from the examination of a 45-year-old female patient with a 5-cm adenocarcinoma. The tumor is clearly visible in the T2-weighted magnitude image of the breast shown in Figure 6a. A 100-Hz wave image is shown in Figure 6b, generated with an electromagnetic actuator and a GRE-based MRE sequence. A very long shear wavelength is observed in the tumor versus the healthy glandular tissues, and the elevated stiffness of the adenocarcinoma is reflected in the elastogram presented in Figure 6c. As seen from the image, the stiff region of the elastogram is consistent with the location observed in the magnitude image.

Brain

Unlike breast tissue, which is easily palpated, brain tissue is not easily accessible except at surgery. Hence, there is significant interest in the potential of MRE to assess the mechanical properties of the brain [66,82,83]. Despite the lack of clinical precedence in some cases, investigators hypothesize that MRE may be useful in the evaluation of certain diseases such as Alzheimer's, hydrocephalus, multiple sclerosis and brain cancer. In addition, transient MRE has been used to study the mechanics of traumatic brain injury [84].

Figure 7 presents MRE data from a healthy volunteer, where a pressure-activated driver was used to introduce 60-Hz shear waves into the brain. In Figures 7a and 7b, x- and y-components of the resulting curl-filtered wave data (processed to remove long-wavelength longitudinal waves) are shown. The magnitude image and corresponding elastogram are shown in Figures 7c and 7d, respectively, demonstrating that quantitative measurements of brain stiffness are possible with MRE. Ongoing investigations in this area will likely determine the viability of MRE to assess specific diseases of the brain, such as the diffuse changes in neural tissue due to aging, stroke, multiple sclerosis and the accumulation of amyloid plaques in Alzheimer's disease [XX refs].

Skeletal Muscle

Skeletal muscle is an example of one tissue that undergoes a significant change in stiffness depending on its physiological state [85]. MRE has been used to study the stiffness of normal and diseased skeletal muscle as a function of contractile state [61,70,86]. The presence of neuromuscular dysfunction, for example, has been shown to affect muscle

stiffness, as determined with MRE [87], and MRE has also been used to study the nature of myofascial taught bands [88].

A sagittal magnitude image from the lower extremity of a healthy volunteer is shown in Figure 8a, with labels indicating the locations of the tibialis anterior, soleus and gastrocnemius muscles. In Figure 8b, a 100-Hz wave image of the unloaded extremity at rest is presented, showing shear wave propagation in all muscles. In Figures 8c and 8d, the subject undergoes resisted plantar and dorsiflexion, respectively, and wavelengths in the corresponding muscle groups are seen to increase in both cases versus the unloaded case. Unlike other tissues, such as the liver, which can readily be considered to behave isotropically, the anisotropic structure (fiber orientation) of skeletal muscle presents special inversion challenges that prevent the use of conventional inversion algorithms and therefore require special consideration for analysis.

Bounded Tissues

In the examples presented thus far, the relative size of the anatomy tends to exceed that of the shear wavelength, meaning that interactions between the propagating shear waves and the tissue boundaries are minimal and can be largely ignored. However, in the instance of some tissues, where the shear wavelength (for practical frequencies) approaches or exceeds the scale of the anatomy, such as the vascular wall, ventricles of the heart, and the corneoscleral shell of the eye, wave propagation becomes severely affected by the presence of these boundary conditions and must be accounted for in the context of MRE. Like other tissues, the mechanical properties of these bounded tissues are also of central importance in a variety of related pathologies, such as hypertension (vascular wall), diastolic dysfunction (heart), and age-related macular degeneration (eye), and are therefore of significant interest from a clinical standpoint and currently under investigation with MRE [37,65,89].

Figure 9 contains magnitude (top row) and wave images (bottom row) of select bounded tissues, including an ex vivo porcine aorta, porcine heart, and bovine globe. In Figure 9d, 100-Hz flexural waves in the vascular wall are detected as corresponding displacements in the intraluminal fluid, observed to be considerably longer-wavelength than the thickness of the aortic wall. In Figure 9e, 200-Hz flexural waves are directly observed in the wall of the left ventricle, guided by the presence of the ventricular boundaries. Similar to the vascular wall example, in Figure 9f, 300-Hz flexural waves in the corneoscleral shell are also observed as displacements in the fluid-filled intraocular space. In this particular example two distinct wavelengths are observed in tissues with markedly different elastic properties. Wavelengths in the cornea (top) are considerably shorter than those in the sclera (bottom), an indication of the cornea's lower stiffness.

Images of waves propagating through bounded and/or pressurized tissues have been processed using a variety of methods (to calculate material properties), such as analytical models of thin-walled tubes [89] and spherical shells [90], and numerical approaches, such as finite element analysis, based on accurate models of the tissue geometry [37,58]. Also of note in these three examples, is that in all cases, the tissue of interest is loaded by a

pressurized fluid, which presents an additional confounding factor, and further opportunity for investigation.

Future Directions

Many challenges and opportunities exist for further development of MRE and continued efforts to explore and characterize the relationship between tissue mechanics and pathology. Ongoing development efforts include various methods for improving the capability of MRE to image propagating shear waves. Driver technology, for example, continues to evolve in pursuit of improved shear-wave illumination at higher frequencies for specific anatomical applications. Phased arrays of multiple drivers, in particular, are currently under investigation for improving the uniformity of shear wave illumination [91]. The ability of most conventional whole-body MRI scanners to perform high-frequency MRE on very stiff tissues such as bone and cartilage is currently limited by gradient hardware performance, a problem that may be addressed, in part, by the design of specialized motion encoding gradient inserts [67]. In addition, advanced pulse sequences for imaging shear waves are also under development in an effort to improve the speed of MRE acquisitions, and to provide high-quality, three-dimensional wave data that includes all three components of motion [49,92]. Finally, the mathematical techniques for analyzing wave images have also continued to evolve in an effort to improve tissue mechanical characterization. Ongoing developments in this area include methods to account for more complex tissue behavior including viscoelasticity, anisotropy, non-linearity, and heterogeneity [12,42,72,86,93,94].

Conclusions

Magnetic resonance elastography (MRE) is a rapidly evolving technology that provides an intuitive and highly-useful source of mechanical contrast by noninvasively probing the tissue properties elicited by palpation. MRE has already been established as a highly-accurate clinical tool for the diagnosis of early-stage hepatic fibrosis. As MRE continues to improve and develop, numerous other applications of this technology, such as those discussed in this review article, may become equally viable clinical tools able to provide clinicians and researchers with useful diagnostic information on a broad range of conditions and pathologies.

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Figure 1.
Pressure-activated acoustic driver system.

MRE Pulse Sequence

Figure 2.
Gradient-echo-based MRE pulse sequence.

Figure 3.
Effective shear stiffness for various tissue types as reported with MRE.

Magnitude Image**Wave Image****Elastogram**

Figure 4.
Tissue-mimicking phantom MRE experiment.

Figure 5.
MRE of the liver.

Figure 6.
MRE of the breast.

Figure 7.
MRE of the brain.

Figure 8.
MRE of skeletal muscle.

Figure 9.
MRE of select pressurized tissues.