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MR Elastography of the Breast: Evolution of Technique, Case Examples, and Future Directions

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Abstract

Recognizing that breast cancers present as firm, stiff lesions, the foundation of breast magnetic resonance elastography (MRE) is to combine tissue stiffness parameters with sensitive breast MR contrast-enhanced imaging. Breast MRE is a non-ionizing, cross-sectional MR imaging technique that provides for quantitative viscoelastic properties, including tissue stiffness, elasticity, and viscosity, of breast tissues. Currently, the technique continues to evolve as research surrounding the use of MRE in breast tissue is still developing. In the setting of a newly diagnosed cancer, associated desmoplasia, stiffening of the surrounding stroma, and necrosis are known to be prognostic factors that can add diagnostic information to patient treatment algorithms. In fact, mechanical properties of the tissue might also influence breast cancer risk. For these reasons, exploration of breast MRE has great clinical value. In this review, we will: (1) address the evolution of the various MRE techniques; (2) provide a brief overview of the current clinical studies in breast MRE with interspersed case examples; and (3) suggest directions for future research.

Keywords

Breast elasticity; Breast imaging; Breast MR imaging; Breast tissue stiffness; Breast viscosity

Introduction

Breast magnetic resonance imaging (MRI) is currently used in both screening and diagnostic settings to detect and characterize breast cancer. Although breast MRI has high sensitivity (90%–100%) for detecting breast cancer, it is limited by a wide range of specificity (37%–97%) because both benign and malignant lesions can enhance when using contrast-enhanced MRI (CE-MRI).^{1–3} Advanced imaging techniques are being studied to improve the specificity of MRI, including diffusion weighted imaging, apparent diffusion coefficient mapping, spectroscopic imaging, and elastography.

Breast MR elastography (MRE) is a non-ionizing, noninvasive, cross-sectional MR imaging technique that provides for quantitative viscoelastic properties of breast tissues following application of external stress. Currently, the technique continues to evolve, as research surrounding the use of MRE in breast tissue is still developing. Measuring tissue stiffness leverages the fact that breast carcinomas have been shown to have a high stiffness compared with surrounding benign tissues.⁴ This is one of the underlying principles of why newly palpable masses are concerning for breast radiologists and clinicians. In the setting of a newly diagnosed cancer, associated desmoplasia, stiffening of the surrounding stroma, and necrosis are known to be prognostic factors that can add diagnostic information to patient treatment algorithms.^{5–12} In fact, mechanical properties of the tissue might also influence breast cancer risk. For these reasons, exploration of breast MRE has strong potential clinical value.

Historically, the concept of elastography was first introduced after 1990 in the form of ultrasound elastography.^{13,14} The MRE technique for the breast was initially reported by 2 groups in 1998; Bishop et al, based at the University of Toronto, and Lawrence et al, based at the Mayo Clinic in Rochester, Minnesota.^{15,16} In 2000, it was subsequently examined by Sinkus et al, based at Kings College, London, United Kingdom.¹⁷ The breast MRE technique has evolved over the past 2 decades. Research includes at least 16 original research papers using slightly different variations in technique and post-processing to optimize images, with the largest study to date evaluating 68 breast lesions.^{18,19} In this review, we will: (1) address the evolution of the various MRE techniques; (2) provide a brief overview of the current clinical studies in breast MRE with interspersed case examples; and (3) suggest directions for future research.

Background of Breast MRE

Recognizing that breast cancers present as firm, stiff lesions, the foundation of breast MRE is to combine tissue stiffness parameters with sensitive breast MR contrast-enhanced imaging. The foundation of breast MRE is to study the biomechanical properties of tissues, which have been hypothesized to correlate with breast cancer risk.⁶ At a molecular level, tissue stiffness reflects the complex interactions between cells and the extracellular matrix.²⁰ Breast cancer is associated with matrix remodeling and stiffening of breast stroma, which can enhance epithelial cell growth, breast tissue reorganization, and promote cell invasion and survival.^{9,20} Increases in the number of cells, amount of collagen, and proteoglycan expression have been shown to increase tissue stiffness.^{21,22} It is these microscopic and macroscopic factors that are being studied in vivo with MRE.

Breast ultrasound elastography also aims to capture similar principles of tissue stiffness and has been shown to improve the specificity of ultrasound.^{23–26} In fact, breast ultrasound elastography has been shown to decrease unnecessary biopsies; however, it is limited by inter- and intra-reader variability and lower spatial resolution.²⁷ Ultrasound elastography is easy to use, widely available, portable, and low-cost; however, it lacks quantitative information on tissue elastic properties.^{4,13,28} MR-based elastography, on the other hand, can measure all 3 spatial components of induced tissue displacement with high accuracy, yielding more precise quantitative elastography measurements.^{11,13,29–34}

MRE Technique Overview

There are 3 steps in obtaining a breast MRE image: (1) delivery of low-frequency mechanical vibrations into the breasts; (2) application of an MRE imaging sequence that acquires the wave displacement field in the breasts; and (3) generation of stiffness maps, are referred to as elastograms, which provide quantitative cross-sectional images depicting the stiffness of tissue using specialized software, called an inversion algorithm. These are then interpreted by the radiologist.

Delivery of Shear Waves

To obtain high quality diagnostic images, technologists aim for proper positioning of a passive driver to optimize shear wave penetration to the maximal area of breast tissue. This is combined with a diagnostic radiofrequency coil in the MRE setup. Improper positioning of the breasts and/or the MRE passive driver will cause inadequate delivery of shear waves and may generate artifacts that decrease the sensitivity of breast MRE. Various setups have been proposed to transmit low-amplitude, high-frequency vibrations into breast tissues. In 2000, Sinkus et al demonstrated better wave penetration and hence better quality images with a longitudinal application of the mechanical waves into the breast tissue rather than when applying waves in a transverse direction (shearing).¹⁷ This has resulted in the conventional, preferred method in most current studies.

Currently, there are 2 common approaches to applying longitudinal mechanical waves for MRE exams. The older method is the use of an active driver, which contacts the breast tissues directly and causes breast tissue compression.^{18,35–37} When breast tissue is compressed, however, the tissue stiffness is altered owing to its nonlinear biomechanical property, known as the compression effect.³⁸ Therefore, when vibrations are introduced in the breast, they can be displaced owing to driver contact with the tissue. Additionally, this contact with and compression of the breast tissue can cause artifacts including areas of signal inhomogeneity, anatomical distortions, and poor image quality.²⁷ Compression effect is difficult to correct unless the distribution of breast compression can be measured or controlled accurately, which would be very difficult. Compression can also cause artifacts in the CE-MRI portion of the exam, which has been well-reported in the literature and may effect breast lesion detection, evaluation of lesion size, image interpretation, and decreased perfusion of contrast into the tissue, which may lead to delayed or non-enhancement of breast lesions.^{39,40}

Newer, non-compressive drivers avoid potential compression effect, maintaining a consistent state of pre-stress.²⁷ A comparison between the compressive and non-compressive setup is highlighted in Figure 1. The newer method delivers mechanical waves (MRE active driver) to both breasts via a passive driver positioned between the sternum and the radiofrequency coil,²⁷ seen in Figure 2. In 2012, Hawley et al demonstrated, in 22 patients, that a consistent state of pre-stress is maintained using a sternal driver that enables reproducible measurements of stiffness.²⁷ In these 22 volunteers, stiffness correlated with Breast Imaging-Reporting and Data Systems (BI-RADS) breast parenchyma density.

Optimal Selection of Vibration Frequency

Once introduced, the mechanical shear waves propagate through the breast tissue at different speeds based on tissue stiffness, reflected by their shear wavelength. Tissues with increased stiffness have a faster propagation speed and longer wavelengths, whereas tissues with low stiffness have a slower propagation speed and shorter wavelengths.

A recurrent issue in multiple studies was an overlap in statistics on stiffness of different tissues (for instance, the overlap between soft malignant tumors and stiff benign lesions). Future clinical trials should lead to improving the sharpness and fidelity of MRE to enhance resolution and quantitative accuracy.³⁷ An important consideration for the breast MRE technique is the frequency of the mechanical vibrations. The vibration frequency has important implications for resolution and signal-to-noise of the stiffness map. As frequency increases, the ability to resolve small regions of differing stiffness (eg, tumors) increases. However, the higher the frequency, the more the shear waves will attenuate in tissue and decrease the MRE signal.⁴² Breast MREs have been performed using vibration frequencies between 37.5 Hz and 300 Hz and have been performed on both 1.5T and 3T scanners.¹⁹ An early study by Lawrence et al of 9 volunteers on a 1.5T scanner demonstrated that the 50 to 100 Hz range was the most suitable frequency range for imaging in vivo breast MRE.^{16,35} In general, the optimal frequency will represent a good compromise between efficient wave penetration deep into the breast tissues (low frequencies) and sufficient spatial resolution (high frequencies).^{35–37} Thus, further research will need to explore the potential for technical improvements in MRE to increase the contrast between various lesions and improve sensitivity. Once the waves are transmitted into breast tissue, an MRE imaging sequence is used to acquire the wave displacement field in the breast.

Image Acquisition Technique

MRE imaging is a phase-contrast imaging technique that can be used to capture a harmonic motion, utilizing cyclic motion-encoding gradients (MEG) synchronized to the mechanical waves in the x, y, or z directions, which captures displacements associated with wave propagation.⁴² The MRE acquisition is able to detect motion in the order of a few hundreds of nanometers.⁴³ These cyclic MEGs are incorporated into a variety of pulse sequences, with motion being recorded using 2-dimensional (D) or 3-D gradient echo, spin echo, echo-planar imaging, balanced steady-state free precession, spiral, and stimulated pulse echo sequences.⁴⁴ The mechanical properties are not dependent on magnetic field strength, allowing for good reproducibility across studies.⁴² Image acquisition parameters are outlined in Table 1.

Reported acquisition times for the MRE sequence have varied between 2.1 minutes to 38 minutes. Number of slices have varied between 1 and 50 with a slice thickness of 3 to 10 mm. Long scan times with few slices were noted in older studies (one study reporting scan times of up to 30 minutes to obtain up to 10 slices), but scan times have greatly improved with the newer techniques.^{45,46} To obtain the most diagnostic information, the goal is to obtain whole breast MRE images with a good resolution and scan time. In 2017, Balleyguier et al demonstrated a 10- to 12-minute scan time of obtaining whole-breast MRE covering 50 slices.⁴⁷ Research teams have worked on integration of MRE with currently used MR

techniques, such as using a gradient echo MR sequence and MR mammography.⁴⁶ By implementing MRE imaging methods with the established breast MR clinical protocols, the process of diagnosis and analysis of breast tissue can be made more efficient.⁴⁷ Reproducibility and repeatability studies are necessary across different breast MRE studies.

Generation of Elastogram Maps: Image Data Post-processing

Several steps are involved for post-processing of the acquired images to create elastograms, or quantitative maps of stiffness, in MRE imaging data reconstruction. The post-processing technique has not been standardized across vendors and groups for breast MRE. In the literature, several different reconstruction methods have been reported.

Once the elastograms are generated, regions of interest (ROIs) are manually drawn in the area of interest in the breast tissue. Many breast MRE studies have included both ROI-based and global measurements of breast biomechanical properties. In the study of Siegmann,³⁵ ROIs were set manually by 2 operators in consensus to enclose the corresponding lesion. ROIs were positioned on the post contrast series with the best visibility of the lesion and then copied over to the corresponding elastograms to calculate corresponding quantitative values. Inter- and intra-reader studies are needed across different breast MRE interpretation techniques.

At our institution, 3-D wave images are processed with 3-D multiple-model direct inversions without directional filtering, generating 3-D elastograms of bilateral breasts (Figure 3). ROIs are then drawn over the entire breast or lesion of interest on magnitude maps, and custom, in-house software is used to obtain MRE stiffness values. Some varied representative elastogram maps are demonstrated in Figures 3–5.

Parameters to Measure Tissue Mechanical Properties

Multiple parameters have been studied to measure tissue mechanical properties. Elastic modulus (λ) is the tissue property that quantifies an object's ability to resist deformation and is a measure of rigidity/stiffness.⁴² The bulk modulus (K) defines a material's resistance to compression.⁴² The shear modulus (G or μ) and the Young's modulus (E) represent the resistance of a material to deformation by shearing and extension, respectively. They are proportional to each other by the equation $G = E/3$, which is based on the principle that biological tissue conserves volume under deformation by the Poisson ratio of 0.5.⁴² Shear modulus is preferable to bulk modulus in characterization of tissue, as the velocity of compression waves does not vary much for different tissues in the body, whereas the shear wave velocity does.⁴² Because tissue is viscoelastic and demonstrates reversible deformation and viscous dissipation, and mechanical properties are measured under dynamic conditions, these factors are also considered in evaluation of tissue stiffness. The storage modulus (G') represents the elastic behavior of tissue and its ability to store elastic energy, whereas the loss modulus (G'') measures the tissue's ability to dissipate energy. These are both incorporated into the complex shear modulus (G^*) by the equation $G^* = G' + G''$. MRE reports the magnitude of the complex shear modulus ($|G^*|$),⁴² which links the applied stress to a given strain can be calculated at a given frequency, allowing for dynamic evaluation.¹⁸ A study by Siegmann et al demonstrated that including other viscoelastic

properties could significantly increase the diagnostic accuracy of breast lesions.³⁵ A recent study by Balleyguier et al demonstrated that the phase angle increased diagnostic frequency and was helpful in predicting malignancy, independent of MRI BI-RADS.

Clinical Applications

Stiffness Differences Between Healthy Tissues Quantified by MRE

Preliminary studies have shown promising results for breast MRE to be used as a complementary tool to improve the specificity of breast MRI (Table 2). Dating back to 1998, Lawrence et al evaluated 9 healthy volunteers and demonstrated that there was a 4- to 7-fold difference between the shear modulus of glandular tissue, which was 2.45 kPa (+0.2 kPa), compared with fat, which was 0.43 kPa (+0.07 kPa). In 2002, McKnight et al evaluated 6 healthy volunteers and demonstrated that fibroglandular tissue had a slightly higher mean shear stiffness of 7.5 ± 3.6 kPa compared with adipose tissue, which was 3.3 ± 1.9 kPa.³⁷ Despite the difference in absolute kPa values between the studies, this result demonstrates a difference in biological tissue properties of dense versus non-dense tissues (Figure 4). In 2017, Hawley et al evaluated 22 healthy volunteers on a 3T scanner and demonstrated that dense breast, including heterogenous dense and extremely dense breasts, had a significantly higher mean stiffness value of 0.92 kPa compared with fatty and scattered fibroglandular breasts, which had mean stiffness values of 0.83 kPa ($P = .05$).²⁷ In 2019, Chen et al studied the relationship between tissue stiffness measured on breast ultrasound elastography to the percent breast density measured on breast MR imaging in 20 patients. This group found no correlation between breast stiffness and breast percent density.³⁸ These preliminary studies suggest the stiffness of the breast stroma is the product of complex interactions within the extracellular matrix, not a simple measure of breast density.

Breast MRE also shows changes in normal fibroglandular tissue based on menstrual cycle and breast density. In 2003, a study by Lorenzen et al demonstrated that there was a significant cyclic change in breast tissue elasticity in phase with the menstrual cycle.⁴⁸ Elasticity values were highest between days 11 and 25 (+35%), and lowest on day 5 after onset of menses (−29%).⁴⁸ Albeit these are low numbers, but this is in keeping with our expectations considering the effect of hormonal background on breast tissues during the menstrual cycle. Maximum proliferation of mammary epithelium is documented during the second half of the menstrual cycle, whereas the lowest proliferation is noted during days 5 to 10.^{49–53} Future studies are needed to confirm this relationship and the overall effect in patients with breast tumors and at elevated risk of breast cancer.⁵⁴

Tumor Characterization

Multiple studies have investigated the added value of MRE in differentiation of benign versus malignant breast lesions (Figure 5). Early studies using phantoms demonstrated the ability of MRE to differentiate healthy tissues from breast tumors. At a frequency of 100 Hz, a breast cancer phantom demonstrated a focal area of higher shear stiffness of 11.7 ± 2.0 kPa at the location of the simulated tumor, whereas the gel surrounding the simulated tumor had a shear stiffness of 2.6 ± 0.4 kPa.⁵⁵ In 2002, McKnight et al performed MRE on 3 mastectomy specimens in patients with breast cancer and noted that mean shear stiffness

was approximately 36 ± 5 kPa in all 3 samples, 580% higher than the average shear stiffness of surrounding adipose-fibroglandular tissue.³⁷ Additionally, among 6 patients with breast carcinoma, the measured shear stiffness of the tumors ranged from 18 to 94 kPa (mean, 33 kPa), whereas that of adipose breast tissue in these patients ranged from 4 to 16 kPa (mean, 8 kPa; $P < .05$).³⁷ They concluded that the mean shear stiffness of breast carcinoma was 418% higher than the mean value of surrounding breast tissues.³⁷ In 2002, Lorenzen et al studied 35 patients and concluded that malignant lesions had a median elasticity of 15.9 kPa (range, 8–28 kPa), whereas the benign lesions had a median value of 7 kPa (range, 2–8 kPa; $P = .0012$).⁴⁶ They noted, however, that in healthy volunteers, stiffness in areas of breast parenchyma without lesions ranged up to 15 kPa, overlapping with malignant lesions.⁴⁶ Xydeas et al evaluated 17 breast lesions, 11 benign and 6 malignant, and noted that malignant lesions were stiffer than fibroadenoma ($P < .0004$), fibrocystic changes ($P < .04$), and overall breast tissue excluding the lesion ($P < .0005$).³⁶ Xydeas et al also demonstrated that there was no significant difference in viscosity between malignant lesions and fibroadenomas, fibrocystic changes, and overall breast tissue excluding the lesion.³⁶ Further investigations into the normal stiffness variation in healthy breast tissue may provide insight on the overlapping stiffness ranges of malignant and benign tumors.

Subsequently, studies with larger patient populations have demonstrated that malignant lesions with more aggressive features or certain subtypes may exhibit more “liquid-like behavior.”^{18,35} This results in a decreased scale of attenuation compared with benign lesions, suggesting that viscosity- and/or frequency-dependent power-law behavior of tissue may be necessary to incorporate in the equation to improve discrimination between benign and malignant lesions.¹⁹ Balleyguier et al evaluated 43 lesions and noted that elasticity decreased and phase angle increased with the BI-RADS score, whereas viscosity and phase angle were increased in malignant lesions.⁴⁷ This study demonstrated that phase angle was important in predicting malignancy independently of the BI-RADS score.⁴⁷ Authors conclude that the phase angle, an objective, quantitative measure that is a combination of several viscoelastic properties of lesions, and less sensitive to variations of one parameter, is preferred as an adjunct to BI-RADS score, rather than viscosity. However, following histologic analysis, there was no correlation between viscoelastic parameters and type of stroma, vessels, or histologic grade. A study by Sinkus et al evaluating 68 lesions demonstrated a 20% increase in specificity at 100% sensitivity with the addition of dynamic MRE to CE-MRI of the breast.¹⁸ A subsequent study by Siegmann et al evaluating breast lesions in 57 patients demonstrated that the specificity in their study increased from 75% to 90% at 90% sensitivity.³⁵ Balleyguier also demonstrated that the addition of MRE to CE-MRI significantly increases diagnostic accuracy, with an area under the curve that increased from 0.84 to 0.92 with the addition of MRE.⁴⁷

Macroscopic Stiffness Parameters are Biomarkers of Collagen Deposition

Most recently, in a study that coupled MRE with computational histopathology, Li et al demonstrated in animal models that quantification of elasticity and viscosity are both sensitive imaging biomarkers of tumor collagen deposition and correspond directly to response enzymatic degradation of the collagen network.^{38,56–58} Collagen is known to increase tumor stiffness, regulate tumor immunity, and promote metastases.⁵⁹

Therefore, these collagen-based studies will prove useful as scientists better understand the pathophysiologic functions of collagen in breast cancer. Collagen can be used as a predictor of prognosis and recurrence in cancers, a biomarker for therapy resistance, and a potential targeted-therapy agent. Specifically, there are promising targeted therapeutic options for modifying the tumor matrix and guiding new strategies for anticancer therapies.^{59–61}

Future Directions

Currently, breast MRE is considered to be investigational.¹³ In light of the above findings, however, it is promising future research that could take a forward-looking direction. For example, research results suggest the capacity of MRE not only as diagnosis but for predicting the prognostic features of breast malignancies and monitoring treatment with neoadjuvant chemotherapy.²⁷ As neoadjuvant treatment usage increases, complementary use of new imaging techniques will be critical for assessing in vivo treatment response.^{8,62–64} Studies have found that a decrease in tumor interstitial pressure may be a biomarker of response to therapy^{65–67} and predictive of improved delivery of anti-cancer drugs. Similarly, an absence in response to collagen changes in the setting of neoadjuvant therapy has predicted therapy resistance in cancer cells.^{68,69} Further investigations will assess in vivo tumor response using changes in tissue stiffness, elasticity, and viscosity to denote tumor response and necrosis.^{10,70}

Given the importance of breast density and breast cancer risk, it is a priority to study how (if at all) tissue stiffness varies in women based on breast cancer risk. Forty percent of women over the age of 40 have dense tissue. Dense breast tissue reduces the effectiveness of mammography screening and increases the risk for developing breast cancer. Elevated extracellular matrix stiffness is a prognostic indicator of tumor progression. This leads to the hypothesis that quantitative biomechanical tissue parameters may help providers with risk stratification in women with overall increased breast density. With improvements in the breast MRE technique such as the advanced sternal drivers, researchers are now able to perform whole breast MRE in a reasonable time frame with good reproducibility, allowing us to study whole breast stiffness and whether this varies based on breast density and breast cancer risk. MRE has the potential to inform noninvasively on prognosis for patients with cancer, inform breast cancer risk stratification, and accelerate the development of stromal targeting/modulating treatment strategies.¹⁰

Conclusions

Breast MRE has shown promise in providing insight into studying quantitative mechanical properties of breast tissue, including stiffness, elasticity, and viscosity. Breast MRE is compatible with current breast MRI coils and continues to evolve as an easy-to-integrate tool into an imaging protocol. The technical component of obtaining MRE images continues to evolve in order to: (1) decrease time to scan and (2) improve spatial resolution. Initial clinical studies have demonstrated the added value of breast MRE to improve the specificity of standard breast MR imaging. More work is needed to determine the value of MRE in assessing a patient's risk of breast cancer and its ability to serve as a prognostic biomarker in malignant lesions. Future work will study the effect of the biomechanical environment on

breast cancer and the relationship between breast density and MRE-based breast cancer risk with stiffness.

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Figure 1.

Overview of breast MRE technique. A, To Generate the Mechanical Waves and Transmit Them Into the Breast, a Driver is Used. The Patient Lays on Top of the Breast RF Coil in a Normal Prone Position for Breast MRI Exams. Novel Pneumatic, Non-compressive Breast MRE Setup. 1 = MRI; 2 = Patient in Prone Position; 3 = Breast RF Coil; 4 = Non-compressive MRE Driver; 5 = Patient Table. The Driver (4) is Positioned on the Breast RF Coil where the Patient Sternum Will Rest Enabling Efficient and Consistent Motion Delivery. In This Example, Acoustic Pressure is Transmitted From an Active Driver (Typically Situated in the MRI Equipment Room) to a Passive Driver (Prone Position) (Resoundant, Mayo Clinic, Rochester, MN). In This Setup, There is No Mechanical Compression to the Breast Tissues and Both Breasts can be Imaged Simultaneously Using the Soft Sternal Driver, which May Improve Image Quality and Decrease Imaging Time. B, Non-compressive Breast MRE Driver ($2 \times 0.6 \times 22$ cm) Design. C, 3-D GRE MRE for Breast Imaging. Vibration is Continuous During the Whole Scan⁴¹

Abbreviations: 3-D = 3-dimensional; GRE = gradient echo; MRE = magnetic resonance elastography; MRI = magnetic resonance imaging; RF = radiofrequency; TE = echo time.

Figure 2. Breast MRE example images. A, T1 Weighted Magnitude, In-phase, out of Phase and T2 Weighted Images From a 35-Year-old Healthy Dense-Breasted Volunteer. B, Curl Waves Demonstrate Appropriate Wave Transmission Throughout the Series. C, Lastly, Post-processed Stiffness Maps Demonstrate the Color-Encoded Elastogram Sequences Obtained From a Magnetic Resonance Elastography Acquisition. A Compilation of the Information From the Various Sequences is Used to Create the Elastogram that Reflects the Stiffness in Various Areas of the Breast Tissue

Figure 3. Magnetic Resonance Elastography Images From a Normal Healthy Volunteer. Magnitude (A) and Associated Color (B) Maps Demonstrate Stiffness in Various Parts of the Breast Bilaterally. Stiffness in Adipose Tissue was Between 0.30 and 0.59 kPa, and Stiffness in Fibroglandular Tissue was Between 0.57 and 1.0 kPa

Figure 4.

A 41-Year-old Female With Biopsy-Proven Invasive Ductal Carcinoma, Breast Imaging-Reporting and Data System (BI-RADS) 6. A, Post-contrast Axial Image Demonstrates an Irregular Enhancing Mass (Dashed Arrow) in the Right Retro-areolar Region, Compatible With Biopsy-Proven Malignancy. B, Magnetic Resonance Elastography Stiffness Values Were Calculated Within the Mass, in the Surrounding Fibroglandular Tissue, as Well as the Adipose Tissue. Color Maps Demonstrate Stiffness in Various Parts of the Breast Bilaterally. Stiffness in Adipose Tissue was Measured as 0.41 ± 0.10 kPa, Stiffness in Glandular Tissue was Measured as 0.90 ± 0.18 kPa, and Stiffness in the Invasive Ductal Carcinoma (Solid Arrow) was Measured as 1.42 ± 0.17 kPa

Figure 5.

A 51-Year Old Female With Biopsy-Proven Invasive Ductal Carcinoma in the Inferior Posterior Right Breast. A, Post Contrast Maximum Intensity Projection Demonstrates Lobulated Mass Measuring up to 3.8 cm in Maximal Dimension (Solid Arrow). B, Color Maps Demonstrate Increased Stiffness in the Region of the Known Cancer (Solid Arrow). Patient Underwent 3 months of Neoadjuvant Therapy. C, Post-treatment Post Contrast Maximum Intensity Projection Demonstrating No Suspicious Residual Enhancement in the Right Breast Suggesting Treatment Response (Dashed Arrow). D, Corresponding Color Map Demonstrates Resolution With Minimal Residual Increased Stiffness in the Region of the Initial Biopsy Bed (Dashed Arrow), but Overall Decreased Stiffness in Region of Malignancy. Surgical Pathology From Right Breast Lumpectomy Reported No Residual Tumor, Biopsy Cavity Extends to the Anterior Soft Tissue Margin, and Fibrotic Tumor Bed Extends to the Inferior and Deep Margins

Table 1**Breast Magnetic Resonance Elastography Acquisition Methods**

Author	Year	Scanner Strength, T	Vibration Frequency, Hz	Pulse Sequence	Acquisition Time, minutes	No. Slices	TR/TE	No. Patients (No. lesions)
Bishop	1998	1.5	50	SE	N/R	N/R	300/54	1 (N/A)
Lawrence	1998	1.5	50–200	GRE	12	N/R	40–300/20–60	9 (N/A)
Sinkus	2000	1.5	60	SE	30	N/R	N/R	1 (1)
Plewes	2000	N/R	N/R	STEAM	2.1	1	N/R	1 (N/A)
Lorenzen	2001	1.5	65	SE	30	10	500/45	3 (4)
Lorenzen	2002	1.5	65	SE	30	10	500/45	35 (20)
McKnight	2002	1.5	75–300	2D-GRE	13–40	N/R	100–300/28	12 (6)
Lorenzen	2003	1.5	65	SE	15	7	500/45	5 (N/A)
Van Houten	2003	N/R	100	2D-GRE	38	12	20.6/N/R	5 (N/A)
Sinkus	2005	1.5	65	SE	N/R	N/R	N/R	15 (15)
Sinkus	2005	1.5	65	SE	7	7	500/45	2 (2)
Xydeas	2005	1.5	65	SE	7	7	430/46	20 (17)
Sinkus	2007	1.5	85	SE	7	7	412/47	68 (68)
Siegmann	2010	1.5	85	SE	7	7	412/47	57 (57)
Hawley	2016	3	60	GRE	5	5	25/19.9	22 (N/A)
Balleguier	2018	1.5	37.5	SE-EPI	50	50	412/47	43 (43)

Abbreviations: EPI = echo-planar imaging; GRE = gradient echo; N/A = not available; N/R = not reported; SE = spin echo; STEAM = stimulated echo acquisition mode; TE = echo time; TR = repetition time.

Table 2

Summary of Reported Stiffness Parameters of Normal and Abnormal Breast Tissue

Author	Year	No. Patients (No. lesions)	Vibration Frequency, Hz	Fibroadenoma/Benign Lesions	Benign Fatty Tissue	Benign Fibrog andular Tissue	Malignant Tissue	Reported P Values
Bishop	1998	1 (N/A)	50	N/A	N/A	c: 0.68 ± 0.16 m/s a: 1.6 ± 0.2 nepers/ λ	N/A	N/A
Lawrence	1998	9 (N/A)	50–200	N/A	fat: 0.43 ± 0.07 kPa	FGT: 2.45 ± 0.2 kPa	N/A	N/A
Sinkus	2000	1 (1)	60	N/A	N/A	IEI: $0.5–1$ kPa	IEI: 3.5 kPa	N/A
Plewes	2000	1 (N/A)	N/R	N/A	SNR 16	SNR 10	N/A	N/A
Lorenzen	2001	3 (4)	65	N/A	N/A	N/A	N/A	N/A
Lorenzen	2002	35 (20)	65	Median Gd = 7 kPa; range, $2–8$ kPa	Median elasticity = 2 kPa; range, $0.5–4$ kPa	FGT adjacent to BC: median Gd = 2.5 kPa; range, $1–7$ kPa stiff FGT: median Gd = 7 kPa; range, $3–15$ kPa Volunteer: 7.5 ± 3.6 kPa	BC: median = 15.9 kPa; range, $8–28$ kPa	BC vs. FGT: $P < .001$ BC vs. fatty tissue: $P < .001$ BC vs. benign lesions: $P = .0012$
McKnight	2002	12 (6)	75–300	N/A	Volunteer: 3.3 ± 1.9 kPa Patient: mean = 8 kPa; range, $4–16$ kPa		mean Gd = 33 kPa; range, $18–94$ kPa	BC vs. fatty tissue: $P < .05$
Lorenzen	2003	5 (N/a)	65	N/A	N/A	N/A	N/A	N/A
Van Houten	2003	5 (N/A)	100	N/A	E: $17.1–23.5$ kPa; λ : $41–29$ kPa; ν : $0.35–0.41$	μ : $9.0–10.8$ kPa; E: $24.3–30.3$ kPa; λ : $44–57$ kPa; ν : $0.39–0.40$	N/A	N/A
Sinkus	2005	15 (15)	65	μ 1.3 ± 0.7 kPa; ζ 2.1 ± 1.4 Pa s	N/A	μ 0.87 ± 0.15 kPa; ζ 0.55 ± 0.12 Pa s	μ 2.9 ± 0.3 kPa; ζ 2.4 ± 1.7 Pa s	BC vs. FA: μ ($P < .001$); ζ N/R BC vs. FGT: μ ($P < .001$); ζ ($P < .02$)
Sinkus	2005	2 (2)	65	FA: μ 1.3 ± 0.7 kPa; ζ 2.1 ± 1.4 Pa s mastopathy: μ 1.2 ± 0.4 kPa; ζ 0.8 ± 0.3 Pa s μ 1.4 ± 0.5 kPa; ζ 1.7 ± 0.8 Pa s	N/A	FGT: μ 0.87 ± 0.15 kPa; ζ 0.55 ± 0.12 Pa s	BC: μ 2.9 ± 0.3 kPa; ζ 2.4 ± 1.7 Pa s	BC vs. FA (μ): $P < .001$ BC vs FGT (μ): $P < .00001$
Xydeas	2005	20 (17)	65		N/A	FCC: μ 1.7 ± 0.8 kPa; ζ 1.7 ± 0.9 Pa s FGT: μ 1.2 ± 0.2 kPa; ζ 1.0 ± 0.3 Pa s	μ 3.1 ± 0.7 kPa; ζ 2.1 ± 1.2 Pa s	N/A
Sinkus	2007	68 (68)	85	Gd 23.0 ± 1.2 kPa; GI 3.6 ± 1.2 kPa	Gd 6.6 ± 0.1 kPa; GI 0.30 ± 0.1 kPa	Gd 12.0 ± 0.6 kPa; GI 0.73 ± 0.21 kPa	Gd 17.2 ± 1.7 kPa; GI 3.8 ± 1.1 kPa	N/A
Siegmann	2010	57 (57)	85	N/A	N/A	N/A	N/A	N/A
Hawley	2016	22 (N/A)	60	N/A	N/A	FGT type A: 0.83 ± 0.11 ; FGT type B: 0.85 ± 0.11 kPa; FGT type C: 0.92 ± 0.13 ; FGT type D: 0.96 ± 0.13 kPa	N/A	N/A

Author	Year	No. Patients (No. lesions)	Vibration Frequency, Hz	Fibroadenoma/Benign Lesions	Benign Fatty Tissue	Benign Fibroglandular Tissue	Malignant Tissue	Reported <i>P</i> Values
Balleguier	2018	43 (43)	37.5	GI 0.33 kPa (range = 0.1–0.97 kPa); γ = 0.35 (range = 0.19–0.62)	N/A	N/A	IDC: GI 0.75 kPa (range = 0.28–2.58 kPa); γ = 0.68 (range = 0.47–0.9) ILC: GI 0.57 kPa (range = 0.38–1.04); γ = 0.67 (range = 0.57–0.84)	N/A

Abbreviations: BC = breast cancer; c = shear wave velocity; E = Young's modulus; $|E|$ = isotropic elasticity; FA = fibroadenoma; FGT = fibroglandular tissue; Gd = elasticity; GI = viscosity; N/A = not available; SNR = signal/noise ratio; γ = power law exponent; μ = shear modulus; λ = Lamé modulus; ν = Poisson's ratio; α = attenuation.