

# Influence of Geometric Distortion on Shear Stiffness Measurement in MR Elastography

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## Abstract

**Objective:** MR elastography (MRE) derives tissue shear stiffness from the wavelength of propagating shear waves. Geometric distortion can therefore alter the apparent wavelength and introduce errors in the calculated shear stiffness. This study evaluated how geometric distortion affects shear stiffness measurements in EPI-based MRE, with a focus on the phase-encoding orientation relative to the wave propagation direction and on the influence of vibration frequency. **Materials and Methods:** Phantom experiments were conducted with systematically varied phase-encoding directions and vibration frequencies to characterize distortion effects under controlled conditions. To assess distortion effects in a clinical setting, liver MRE was performed in three healthy volunteers. Distortion correction was applied in both experiments, and distortion effects were quantitatively evaluated by comparing shear stiffness values before and after correction. **Results:** The phantom experiments identified two factors that compromise measurement uniformity. First, the spatial uniformity of shear stiffness measurements decreased when the wave propagation direction was parallel to the phase-encoding direction (*i.e.*, the direction along which distortion occurs). Second, this non-uniformity further increased as the propagating wavelength became shorter, which occurs at higher vibration frequencies or in softer tissue regions. In contrast, liver MRE in healthy volunteers showed minimal distortion effects (mean difference:  $0.10 \pm 0.15$  kPa; 95% CI:  $-0.28$  to  $0.48$  kPa), suggesting that current clinical liver MRE protocols are largely unaffected by geometric distortion. **Conclusion:** As MRE is increasingly applied to organs beyond the liver, geometric distortion may become a more relevant source of error in these emerging applications. Careful consideration of wave propagation patterns and distortion characteristics is therefore essential for accurate and reliable MRE measurements across various organs.

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## Keywords

MR Elastography, Distortion, Echo Planar Imaging, Wave Image, Elastogram

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## 1. Introduction

Magnetic resonance elastography (MRE) is a noninvasive imaging technique that enables quantitative assessment of the mechanical properties of organs, particularly tissue stiffness [1]-[4]. The fundamental principle of MRE involves applying mechanical vibrations to the body surface to generate shear waves within the target tissue and imaging their propagation behavior. The propagating waves are captured as MR phase images using phase-contrast methods with motion encoding gradient (MEG), specialized magnetic gradient fields synchronized with an external mechanical driver. Since shear wave propagation behavior varies with tissue stiffness, shear stiffness can be calculated by analyzing wave characteristics such as displacement amplitude and local wavelength. The accuracy of these calculations relies fundamentally on the precise measurement of these wave parameters. MRE received FDA (Food and Drug Administration) approval for clinical use in the United States in 2009 and is now widely utilized in MRI systems worldwide [5]. Currently, the primary clinical application of MRE is the detection and staging of liver fibrosis [6]-[8]. Furthermore, additional applications of MRE are being developed through research in other organs, including the brain [9]-[12], breast [13]-[15], kidney [16]-[18], pancreas [19]-[21], and skeletal muscle [22]-[27].

Two main types of MRE sequences are commonly used in clinical practice and research: gradient-echo MRE (GRE-MRE) and spin-echo echo-planar imaging MRE (SE-EPI MRE). GRE-MRE is well-established and provides reliable measurements [28] [29]; however, it employs sequential line-by-line k-space acquisition, resulting in longer acquisition times compared to SE-EPI MRE. Accordingly, clinical application of GRE-MRE to study the liver typically requires several temporal steps involving successive breath-holds, which can impose a significant burden on patients [30] [31]. In contrast, SE-EPI MRE allows collection of all k-space data at once following a single radiofrequency (RF) pulse, making it less susceptible to motion artifacts and enabling imaging to be performed within a single breath-hold. However, SE-EPI MRE has a drawback: the long readout duration required for EPI acquisition allows phase differences due to static magnetic field ( $B_0$ ) inhomogeneity and magnetic susceptibility effects to accumulate, resulting in geometric image distortion primarily in the phase-encoding direction [32] [33]. This distortion becomes more severe at higher  $B_0$  field strengths [34]. Because shear stiffness calculations depend directly on accurate measurements of wavelength and displacement amplitude, even modest geometric distortions can alter the apparent values of these wave parameters and lead to clinically significant

measurement errors, potentially resulting in misdiagnosis and inappropriate treatment decisions. Prior work has addressed geometric distortion in EPI-MRE, proposing correction methods based on field mapping techniques or image registration approaches [35] [36]. Despite these efforts, these studies have not systematically investigated how the interplay between distortion direction and wave propagation patterns influences the magnitude of measurement errors in calculated shear stiffness. Specifically, it remains unclear whether distortion-induced errors vary depending on the relative orientation between the phase-encoding direction and the principal direction of wave propagation, or how wavelength (*i.e.*, vibration frequency) modulates these effects. Understanding these relationships is essential for optimizing SE-EPI MRE protocols and interpreting clinical results accurately.

Therefore, the present study aimed to systematize the impact of geometric distortion on shear stiffness calculations in SE-EPI MRE. Using phantom experiments with systematically varied wave propagation directions and vibration frequencies, we tested two primary hypotheses: first, that distortion-induced measurement errors would be greatest when the phase-encoding direction aligns parallel to the principal direction of wave propagation; and second, that errors would increase at shorter wavelengths (higher vibration frequencies). To assess the clinical relevance of these findings, we also evaluated geometric distortion effects in liver MRE examinations performed according to established clinical protocols.

## 2. Materials and Methods

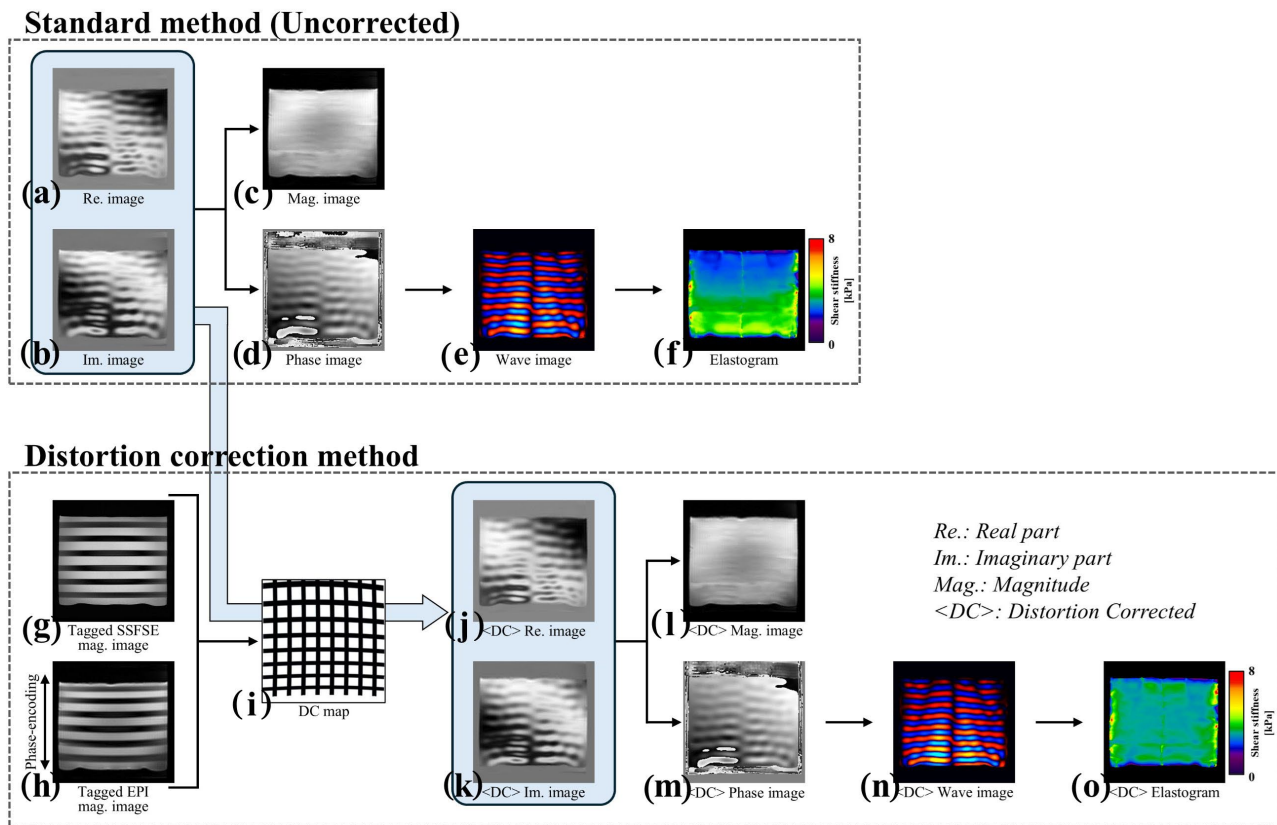
### 2.1. Distortion Correction

In the present study, distortion correction was applied to SE-EPI MRE data to evaluate the effects of geometric distortion on apparent wave parameters and calculated shear stiffness by comparing images before and after correction. For this purpose, we developed a simple custom distortion correction method using a magnitude image with substantially reduced geometric distortion as a reference. The correction process, illustrated in the flowchart in **Figure 1**, consists of the following three main steps:

#### (i) *Acquisition of Reference Images*

Two MR magnitude images were acquired in the absence of any vibrations being induced in the test object. The first magnitude image was acquired using a single-shot fast spin-echo (SSFSE) sequence and the second magnitude image was acquired using the SE-EPI MRE sequence. The SSFSE magnitude image is used as a reference image for calculating distortion in the SE-EPI MRE image. For both magnitude images, multiple equally spaced saturation pulses were applied, producing the so-called tags illustrated in **Figure 1(g)** and **Figure 1(h)**. The degree of curvature and misalignment of the tags serves as a visual marker of geometric distortion effects, which occur principally in the phase-encoding direction. In particular, the tags are applied in the readout direction and appear as lines in the image, making them well-suited to capturing distortions along

the orthogonal phase-encoding direction. A total of six tags, which were available as default options on the MRI system, were applied. Since the geometric distortion of EPI images varies depending on the acquisition parameters that are used, the magnitude images used for the calculation of distortion information were acquired using the same imaging parameters, other than those which refer to the tags, as for the SE-EPI MRE sequence. All MRE images were acquired with the same acquisition parameters in the same orientation and position and with the same field of view (FOV).



**Figure 1.** Process of elastogram calculation for the standard (uncorrected) method and the distortion correction method. In the uncorrected method, magnitude and phase images (c) (d) are calculated from the output real part and imaginary part images (a), (b). The phase image is converted to a wave image (e), and the elastogram (f) is calculated. In the distortion correction method, two additional magnitude images are acquired by single-shot fast spin-echo (SSFSE) sequence (g) and spin-echo-type echo-planar MR elastography (SE-EPI MRE) sequence (h), with multiple saturation pulses (tags) applied to each. Both magnitude images are compared, and a distortion correction (DC) map (i) of the EPI image is calculated. By applying the DC map to the real part and imaginary part images, distortion-corrected real part and imaginary part images (j), (k) are obtained, from which distortion-corrected magnitude and phase images (l), (m) can be created. Using these corrected images, a more accurate wave image (n) and elastogram (o) with reduced image distortion can be obtained.

#### (ii) Calculation of Distortion Information

The tagged SE-EPI MRE magnitude image was compared to the tagged SSFSE reference magnitude image, and a distortion correction (DC) map was produced (Figure 1(i)). This DC map was then applied to correct for distortion in images acquired using the SE-EPI MRE sequence, as described below.

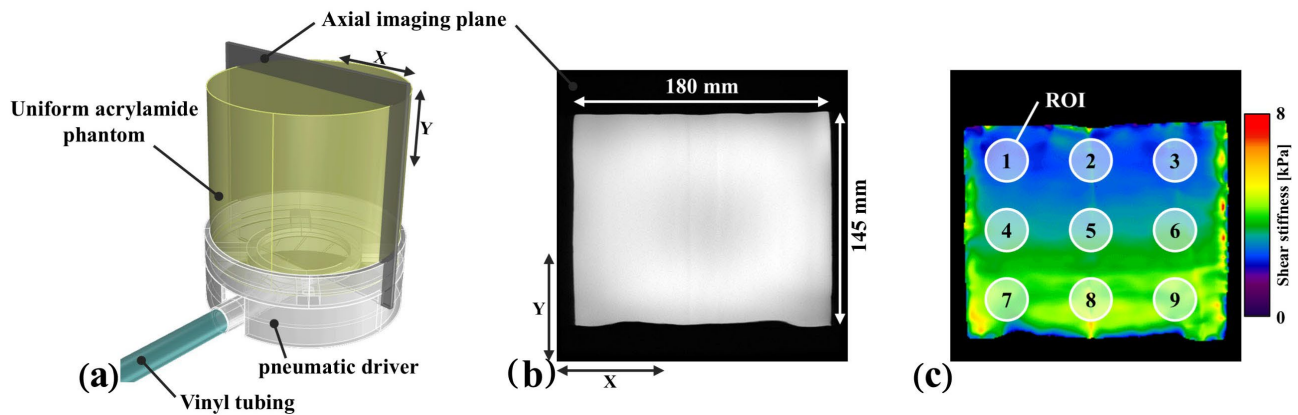
### (iii) *Application of the Distortion Correction*

Since SE-EPI MRE images and the tagged SE-EPI MRE magnitude image acquired without vibrations (**Figure 1(h)**) used for calculation of distortion information were acquired with the same acquisition parameters, except relating specifically to the application of tags, they should have similar geometric distortion. As described by Fehner et al (2017) [35], the DC map obtained in (ii) must be applied separately to the real and imaginary images (**Figure 1(a)** and **Figure 1(b)**), which are then recombined to produce distortion-corrected real and imaginary images (**Figure 1(j)** and **Figure 1(k)**). From these, distortion-corrected magnitude and phase images (**Figure 1(l)** and **Figure 1(m)**) can be produced. This allows generation of more accurate wave images (**Figure 1(n)**) and elastograms (**Figure 1(o)**) with reduced geometric distortion.

## 2.2. Imaging

All experiments were performed using a SIGNA Premier 3 T MRI system (General Electric Healthcare, Milwaukee, WI, United States) equipped with 60 channel posterior phased-array coil and 30 channel anterior phased-array coil.

In the phantom MRE experiment, we used a uniform acrylamide phantom that has been developed for the purposes of quality control in MRE imaging (The Japanese Society for Magnetic Resonance in Medicine, Minato-ku, Tokyo, Japan) (**Figure 2(a)** and **Figure 2(b)**). MRE was performed using a 2D SE-EPI MRE sequence with repetition time (TR) 1,000 ms, echo time (TE) 60.6 to 86.7 ms, temporal phases 4, matrix  $96 \times 96$ , FOV  $20 \text{ cm} \times 20 \text{ cm}$ , slice thickness 8 mm. External vibrations were generated using a TIT320C-4 12" subwoofer (Dayton Audio, Springboro, OH, United States) as a pneumatic pressure generator and were propagated using vinyl tubing to a pneumatic driver placed under the test object so that waves propagated upward toward the top of the phantom (**Figure 2(a)**). The pneumatic driver was designed using Rhinoceros 3D 5.0 modeling software (Robert McNeel and Associates, Seattle, WA, United States) and was fabricated using a 3D printer (3D Systems, Rock Hill, SC, United States). The direction perpendicular to the propagating wave was designated as X, and the direction parallel to the propagating wave was designated as Y. To systematically evaluate how the relative orientation between the phase-encoding direction and wave propagation affects apparent wave parameters, acquisitions were performed with the phase-encoding direction oriented either perpendicular (along X) or parallel (along Y) to the principal direction of wave propagation. MEG was consistently applied along the X-axis throughout phantom experiments. For each orientation, three different vibration frequencies (50, 100, and 150 Hz) were used. Imaging was performed in the axial plane (**Figure 2(b)**). In addition, at each vibration frequency, a reference image to be used in applying the distortion correction (see Section 2.1 (i) above) was acquired using an SSFSE sequence, with TR 1304 ms, TE, 66.8 ms and number of saturation pulses 6 and with the same FOV  $20 \text{ cm} \times 20 \text{ cm}$ , slice thickness 8 mm and positioning as for MRE imaging.



**Figure 2.** Phantom MRE experiment setup and region of interest (ROI) setting in elastograms. (a) Setup diagram of phantom MRE experiments. The phantom is cylindrical with a diameter of 180 mm and a height of 145 mm, composed of homogeneous acrylamide. (b) Axial imaging section. (c) Nine ROI locations in the elastogram for measuring shear stiffness.

In liver MRE, three healthy volunteers (all male, age range: 21 - 24 years) were enrolled in this study. All procedures were performed after obtaining informed consent from the participants and receiving approval from the Institutional Ethics Review Board (approval number: 21045). The acquisition parameters for 2D SE-EPI MRE were set according to the conditions described in Quantitative Imaging Biomarkers Alliance (QIBA) guidelines [37] as follows: TR 1,000 ms, TE 61.4 ms, temporal phases 4, matrix  $96 \times 96$ , FOV  $42 \text{ cm} \times 42 \text{ cm}$ , slice thickness 8 mm, vibration frequency 60 Hz. As in the phantom experiment, external vibrations were applied to the liver using a TIT320C-4 12" subwoofer as a pneumatic pressure generator connected to a passive driver from a commercial MRE system (Resoundant, Rochester, MN, United States). Passive driver placement followed QIBA guidelines. The acquisition parameters for the reference images obtained using SSFSE were as follows: TR 1,041 ms, TE 59.7 ms, number of saturation pulses 6, with the same FOV  $42 \text{ cm} \times 42 \text{ cm}$ , slice thickness 8 mm and positioning as for MRE imaging. The SSFSE reference image and the SE-EPI MRE image were acquired sequentially within the same examination session, each within a single end-expiration breath-hold. Identical breath-hold instructions were provided for both acquisitions to match the respiratory position of the liver and minimize inter-scan displacement.

### 2.3. Image Processing

For distortion correction, the magnitude images produced using the SSFSE and SE-EPI MRE sequences were co-registered using a non-rigid registration method based on B-spline deformation fields [38]-[40] implemented with code written in Python 3.10.9 (Python Software Foundation, Wilmington, DE, United States) and the SimpleITK Ver. 2.2.1 registration framework (Insight Software Consortium, NY, United States). In an iterative approach, the geometric deformation of the SE-EPI MRE image with respect to the SSFSE reference image is computed and corrected by optimizing the spatial correspondence between the two images. Mattes



mutual information [41] was adopted as the similarity metric, which quantifies the statistical dependence between images and provides robust similarity evaluation even between images with different contrast characteristics. The Limited-memory Broyden-Fletcher-Goldfarb-Shanno Bounded (L-BFGS-B) algorithm [42] [43] was used as the optimization algorithm, achieving efficient optimization through quasi-Newton approximation of the Hessian matrix. The deformation model employed B-spline deformation fields using cubic B-spline basis functions, and local and smooth deformations are produced through linear combinations of basis functions defined on a control point grid. Parameters such as control point spacing and convergence criteria were optimized according to the characteristics of the target images. The initial deformation was set as an identity transformation, and deformation parameters were adjusted stepwise during the optimization process. A multi-level optimization strategy was adopted to improve computational efficiency and convergence stability. At the first level, reduction by a factor of 4 and Gaussian smoothing were applied to roughly estimate global deformations, at the second level, a reduction by a factor of 2 was used to refine moderate deformations, and at the third level, fine deformations were finally adjusted at full resolution. The maximum number of L-BFGS-B iterations was set to 200 per level, and the sampling fraction for mutual information estimation was set to 1%. Linear interpolation was used throughout and the flexibility of B-spline basis functions enables handling of complex local deformations. A full 2D B-spline warp was permitted without explicit constraint to the phase-encoding axis. Inspection of the DC maps confirmed that the resulting deformation was predominantly along the phase-encoding direction, consistent with the directional nature of EPI distortion. Finally, a DC map representing the geometric deformation field was generated. The DC map was then applied to the real and imaginary components of the MRE dataset to produce distortion-corrected magnitude and phase images.

The MRE data before and after distortion correction were further processed using MRE-rTool (Keio University Hospital, Shinjuku-ku, Tokyo, Japan) [44]. Elastograms were generated using local frequency estimation (LFE), a method in which shear stiffness is calculated based on local wavelengths [44]-[46]. This approach is particularly sensitive to distortion-induced alterations in apparent wavelength, making it well-suited for evaluating the impact of geometric distortion on shear stiffness quantification.

## 2.4. Data Analysis

The acrylamide phantom has a uniform internal structure with homogeneous mechanical properties. In the absence of geometric distortion, propagating waves should exhibit consistent wavelength, resulting in a uniform elastogram. However, geometric distortion can alter the apparent wave characteristics and lead to artificial variations in calculated shear stiffness. To evaluate this effect, nine circular ROIs (each approximately  $440 \text{ mm}^2$ ) were placed on the elastogram in a  $3 \times 3$  grid pattern, positioned at least 10 mm from the phantom edge to avoid wave

reflection and interference artifacts (**Figure 2(c)**). ROI placement was performed manually by a single operator (Y.I.) on both the uncorrected and corrected elastograms, with ROIs positioned at spatially corresponding locations within the phantom. The coefficient of variation (CV) of the measurements obtained for the nine ROIs was calculated before and after application of the distortion correction, and compared. The Brown-Forsythe test was used to assess whether the variance of shear stiffness values across the nine ROIs differed significantly between the uncorrected and corrected conditions. This test was selected because the principal effect of geometric distortion in a homogeneous phantom is to introduce spatial non-uniformity (*i.e.*, increased variance across ROIs) rather than a systematic shift in mean stiffness. Accordingly, the reported p-values correspond to differences in variance (spatial uniformity), not differences in mean stiffness. Statistical significance was set at  $p < 0.05$ . Because no independent stiffness reference was obtained for the phantom, the phantom endpoint reflects distortion-related spatial non-uniformity rather than absolute accuracy against a ground-truth value.

For liver MRE, a single freeform ROI was manually drawn within the right hepatic lobe on both the uncorrected and corrected elastograms by a single operator (Y.I.), avoiding large vessels, biliary structures, and the liver edge, following QIBA recommendations [37], with ROIs positioned at anatomically corresponding regions. Given the small sample size ( $n = 3$ ), formal statistical hypothesis testing was not performed. Instead, descriptive statistics and 95% confidence intervals for stiffness changes (corrected minus uncorrected values) were calculated using the t-distribution, with results interpreted descriptively in comparison with phantom study findings.

### 3. Results

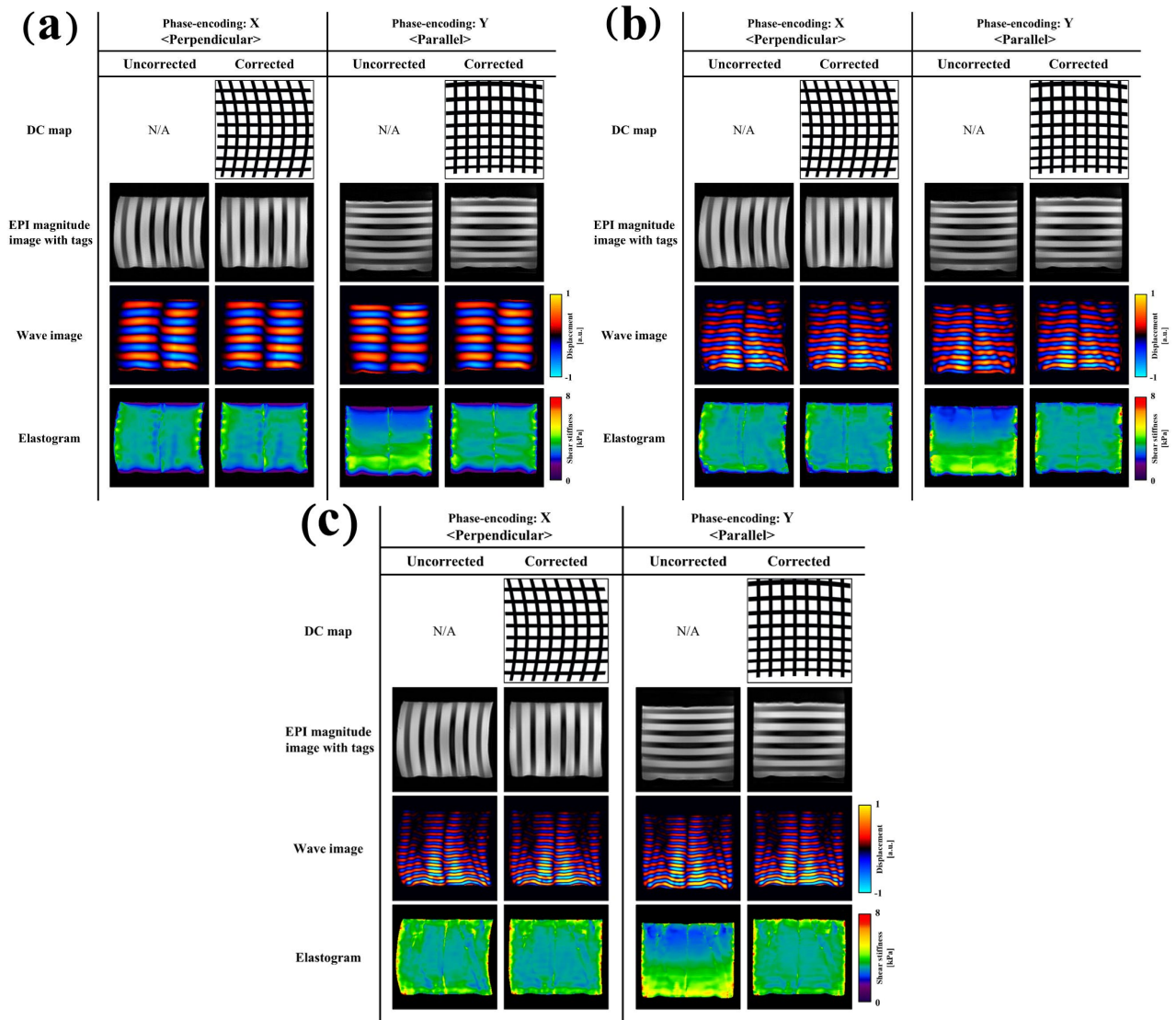
#### 3.1. Phantom MRE Experiment

Tagged EPI magnitude images, wave images, and elastograms before and after distortion correction at vibration frequencies of 50 Hz, 100 Hz, and 150 Hz are shown in **Figure 3(a)-(c)**, respectively. Mean Shear stiffness values and CV values across the nine ROIs for both phase-encoding directions are presented in **Table 1**. **Figure 4** shows stiffness values before and after distortion correction when the phase-encoding direction was perpendicular (a)-(c) and parallel (d)-(f) to the wave propagation direction at each frequency (50 Hz, 100 Hz, and 150 Hz).

Considering first the data acquired at a vibration frequency of 50 Hz, when phase-encoding was applied perpendicular to the direction of wave propagation, obvious distortion was present in the tagged magnitude images (**Figure 3(a)**, left panel, row 2). This distortion was removed by distortion correction (right column). However, because the distortion occurred perpendicular to the direction of wave propagation, it did not affect the apparent wavelength of the propagating waves (row 3). Accordingly, uniformity was comparable in the elastograms obtained before and after distortion correction (row 4), and the CV values were comparable before and after distortion correction. There was no significant difference



in the shear stiffness values within the nine ROIs before and after distortion correction (**Figure 4(a)**;  $p = 0.56$ ).



**Figure 3.** Phantom MRE results: EPI magnitude images with tags, DC maps, wave images, and elastograms at 50 Hz (a), 100 Hz (b), and 150 Hz (c). Left and right panels show phase-encoding gradients applied perpendicular and parallel to the wave propagation direction, respectively. Left and right columns show the results before and after distortion correction. Each panel displays DC maps (row 1, right column only), tagged SE-EPI images (row 2), wave images (row 3), and elastograms (row 4).

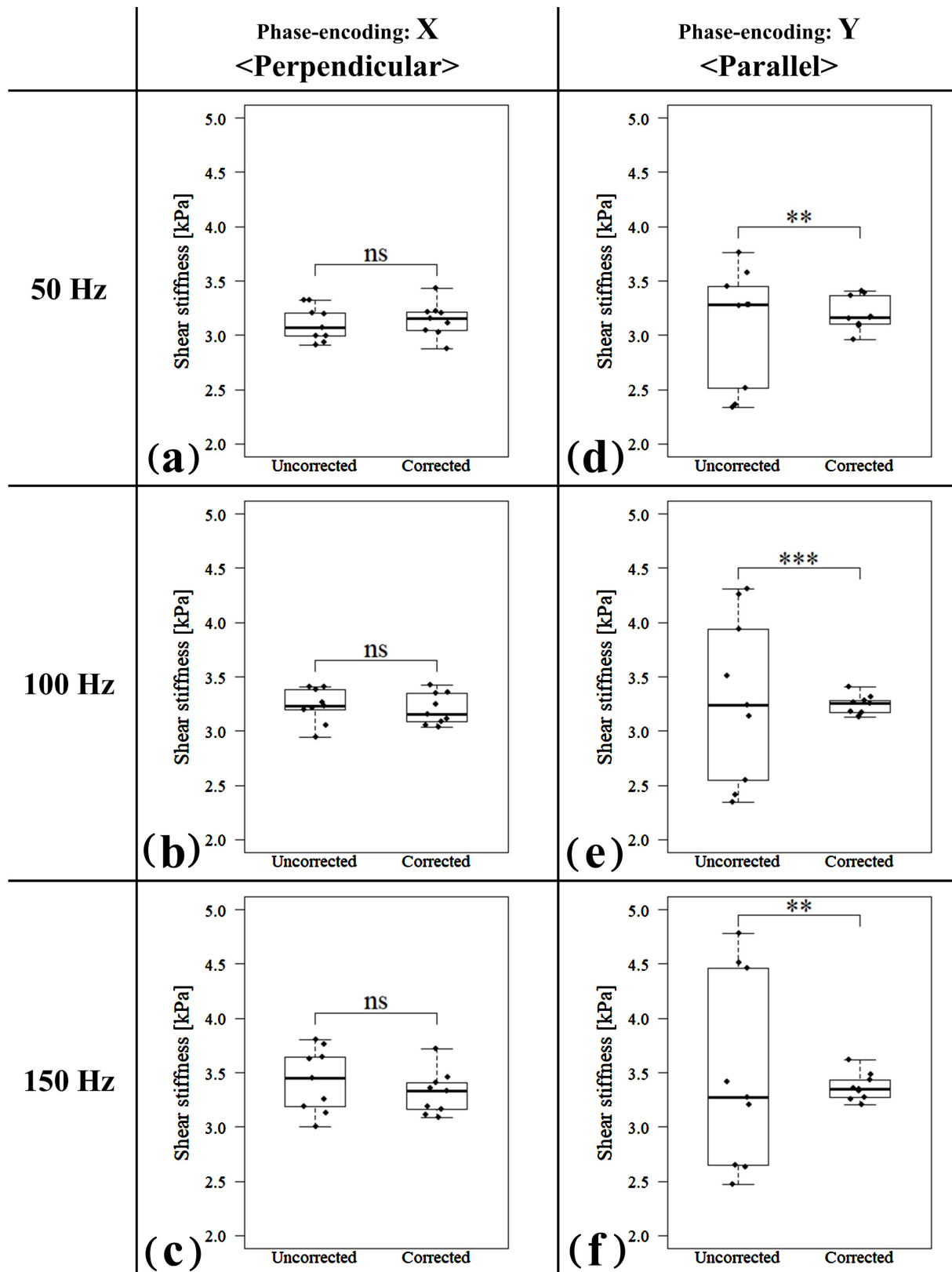
When phase-encoding was in the direction parallel to wave propagation, at first glance the tagged magnitude images appeared comparable before and after distortion correction (**Figure 3(a)**, right panel, row 2). However, detailed examination (including analysis of profiles, not illustrated) revealed alterations in tag spacing in the non-corrected data, with compression in the upper portion and expansion in the lower portion of the phantom. Because this distortion occurred in the direction of wave propagation, it altered the apparent wavelength, resulting in reduced apparent wavelength in the upper portion and increased apparent wave-

length in the lower portion of the image (row 3), producing non-uniform elastograms (row 4). Distortion correction successfully restored equal tag spacing (right column, row 2), consistent apparent wavelength (right column, row 3), and uniform elastograms (right column, row 4). The CV values were lower for the distortion-corrected elastograms, indicating greater improvement compared to when the phase-encoding direction was perpendicular to the direction of wave propagation. There was a significant difference in the variance of shear stiffness values across the nine ROIs before and after distortion correction (**Figure 4(d)**;  $p < 0.01$ ), indicating a reduction in distortion-related spatial non-uniformity.

With regard to the effect of increasing vibration frequency to 100 Hz and then to 150 Hz, when phase-encoding was perpendicular to the direction of wave propagation, there were no apparent differences in the uniformity of the elastograms before and after distortion correction (**Figure 3(b)** and **Figure 3(c)**, left panels). The CV values remained largely unchanged, and there was no significant difference in the stiffness values before and after distortion correction (**Figure 4(b)**; 100 Hz,  $p = 0.94$  and **Figure 4(c)**; 150 Hz,  $p = 0.11$ ). However, when the phase-encoding direction was parallel to the direction of wave propagation, non-uniformity increased with increasing vibration frequency (*i.e.*, with shortening wavelength) in the non-corrected wave images (**Figure 3(b)** and **Figure 3(c)**, right panels, row 3) and elastograms (row 4), and the CV values also increased. Distortion correction successfully restored consistent apparent wavelength (**Figure 3(b)** and **Figure 3(c)**, right panels, row 3, right column) and uniform elastograms (row 4, right column) regardless of vibration frequency, with lower CV values. There were significant differences in the variance of shear stiffness values before and after distortion correction (**Figure 4(e)**; 100 Hz,  $p < 0.001$  and **Figure 4(f)**; 150 Hz,  $p < 0.01$ ), reflecting reduced spatial non-uniformity after correction.

**Table 1.** Mean shear stiffness and coefficient of variation (CV) before and after distortion correction in the phantom experiment.

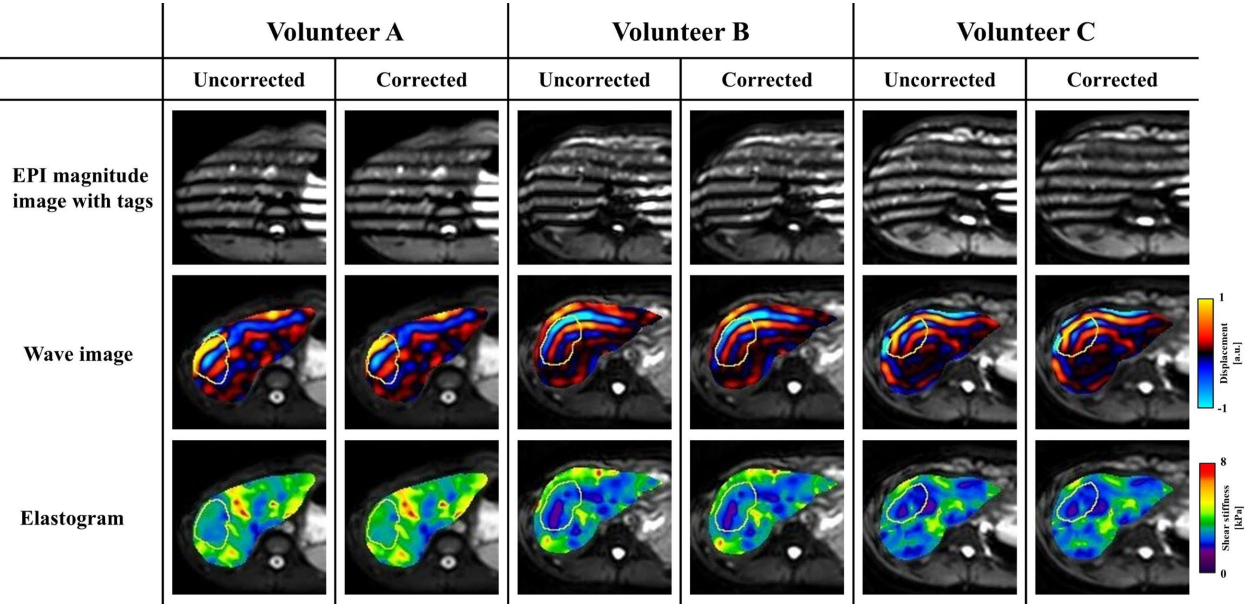
| Vibration frequency [Hz] | Phase-encoding    | Condition   | Mean $\pm$ SD [kPa] | CV [%] | $p$ -value  |
|--------------------------|-------------------|-------------|---------------------|--------|-------------|
| 50                       | X (Perpendicular) | Uncorrected | $3.11 \pm 0.16$     | 5.20   | $p = 0.56$  |
|                          |                   | Corrected   | $3.14 \pm 0.16$     | 4.97   |             |
|                          | Y (Parallel)      | Uncorrected | $3.09 \pm 0.54$     | 17.58  | $p < 0.01$  |
|                          |                   | Corrected   | $3.20 \pm 0.15$     | 4.83   |             |
| 100                      | X (Perpendicular) | Uncorrected | $3.23 \pm 0.16$     | 4.93   | $p = 0.94$  |
|                          |                   | Corrected   | $3.20 \pm 0.15$     | 4.53   |             |
|                          | Y (Parallel)      | Uncorrected | $3.30 \pm 0.77$     | 23.22  | $p < 0.001$ |
|                          |                   | Corrected   | $3.24 \pm 0.09$     | 2.82   |             |
| 150                      | X (Perpendicular) | Uncorrected | $3.43 \pm 0.30$     | 8.63   | $p = 0.11$  |
|                          |                   | Corrected   | $3.31 \pm 0.20$     | 6.09   |             |
|                          | Y (Parallel)      | Uncorrected | $3.49 \pm 0.89$     | 25.47  | $p < 0.01$  |
|                          |                   | Corrected   | $3.37 \pm 0.13$     | 3.79   |             |



**Figure 4.** Phantom MRE results: comparison of stiffness values before and after distortion correction. Phase-encoding direction perpendicular (a)-(c) and parallel (d)-(f) to wave propagation direction at 50 Hz (a) (d), 100 Hz (b) (e), and 150 Hz (c) (f). ns, not significant; \*\*,  $p < 0.01$ ; \*\*\*,  $p < 0.001$ . Brown-Forsythe test for equality of variance across the nine ROIs.

3.2. Liver MRE Experiment

Tagged EPI magnitude images, wave images, and elastograms for all participants before and after distortion correction are shown in **Figure 5**. Shear stiffness values for each volunteer are presented in **Table 2**. The shear stiffness values showed small changes following distortion correction across all three volunteers. The mean difference in stiffness values (corrected minus uncorrected) was  $0.10 \pm 0.15$  kPa, with a 95% confidence interval of  $[-0.28, 0.48]$  kPa. Although this confidence interval included zero, the limited sample size precludes firm conclusions regarding the statistical significance of distortion correction effects *in vivo*.



**Figure 5.** Liver MRE results: EPI magnitude images with tags, wave images, and elastograms of all participants before and after distortion correction. Wave images and elastograms are overlaid on the magnitude images for individual volunteers.

**Table 2.** Shear stiffness of the ROI for all participants in the liver MRE experiment. 95% CI, 95% confidence interval.

| Volunteer | Uncorrected [kPa] | Corrected [kPa] | Difference [kPa] |
|-----------|-------------------|-----------------|------------------|
| A         | 2.99              | 3.26            | +0.27            |
| B         | 2.49              | 2.55            | +0.06            |
| C         | 2.35              | 2.32            | −0.03            |
| Mean ± SD | 2.61 ± 0.34       | 2.71 ± 0.49     | 0.10 ± 0.15      |
| 95% CI    | [1.77, 3.45]      | [1.49, 3.93]    | [−0.28, 0.48]    |

4. Discussion

The phantom experiments validated both hypotheses proposed in this study: first, that geometric distortion most significantly degrades the spatial uniformity of shear stiffness measurements when the phase-encoding direction is oriented parallel to the principal direction of wave propagation; and second, that these effects

become more pronounced at shorter wavelengths (higher vibration frequencies). The underlying mechanism can be explained by the directional nature of geometric distortion in EPI sequences. Specifically, when phase-encoding is parallel to the direction of wave propagation, geometric distortion appears along the wave propagation axis, thereby altering the apparent wavelength of the propagating waves. This alteration in apparent wavelength directly compromises the accuracy of shear stiffness calculation, as MRE relies fundamentally on precise wavelength measurements. Furthermore, this effect becomes greater as the propagating wavelength decreases, even if the degree of geometric distortion remains constant. At shorter wavelengths, a given magnitude of geometric distortion represents a larger fraction of the wavelength itself, thereby producing proportionally greater changes in apparent wavelength and consequently more substantial errors in shear stiffness calculation. These principles are exemplified at 150 Hz compared to 50 Hz. At the higher frequency, shear stiffness decreased substantially more in the upper phantom region, where apparent wavelengths were shortened due to spatial compression, while shear stiffness increased substantially more in the lower region, where apparent wavelengths were lengthened due to spatial expansion. Consequently, despite the uniform phantom, elastograms exhibited increased non-uniformity at higher vibration frequencies as an artifact of geometric distortion. Distortion correction successfully restored consistent apparent wavelengths and uniform elastograms at all vibration frequencies, with convergent shear stiffness values and reduced CV values. Conversely, when phase-encoding was perpendicular to the direction of wave propagation, geometric distortion occurred in a direction transverse to wave propagation and therefore did not alter the apparent wavelength. Consequently, calculated shear stiffness values remained unaffected, and uniform elastograms were consistently obtained at all vibration frequencies, with no significant differences observed before and after distortion correction.

In the liver MRE experiments, geometric distortion was minimal across all participants, with minimal differences between elastograms before and after correction. This contrasts sharply with the phantom results, where distortion effects were substantial. We attribute this difference to three potential factors. First, the clinical MRI system was optimized for human body imaging with acquisition parameters tuned to minimize geometric distortion. Second, the 60 Hz frequency produced longer wavelengths, reducing the proportional impact of geometric distortion on apparent wavelength. Third, wave propagation in liver MRE differed fundamentally from the controlled unidirectional propagation in phantom experiments: waves propagated concentrically, creating a multidirectional wave within the measurement region. Consequently, only a limited portion of the measurement region experienced the parallel alignment between wave propagation and phase-encoding direction that phantom experiments identified as most vulnerable to distortion. Although the sample size and participant characteristics were limited in this study, these results indicate that liver MRE using the QIBA protocol is minimally affected by geometric distortion and achieves relatively high meas-

urement accuracy.

Research is currently underway to expand the application of MRE beyond the liver to other organs. While our liver MRE experiments showed minimal distortion effects, other organs may experience significant geometric distortion due to magnetic susceptibility effects, depending on anatomical location and structural characteristics. This can lead to alterations in apparent wavelength and consequent deterioration in shear stiffness calculation accuracy. Tissues near air-tissue interfaces or those difficult to position at the magnet isocenter experience particularly severe geometric distortion. For such tissues, strategic optimization of both actuator placement and phase-encoding orientation can help minimize distortion effects. By carefully selecting actuator position, wave propagation patterns can be controlled to reduce alignment with the phase-encoding direction, thereby mitigating the orientation-dependent errors characterized in our phantom experiments. Based on the findings from our phantom experiments, successful implementation of MRE across various organs requires careful consideration of the relative orientation between wave propagation and phase-encoding direction. Higher vibration frequencies and measurements in softer tissue regions, which result in shorter propagating wavelengths, also warrant particular attention due to their amplified susceptibility to distortion effects. Attention to these factors is essential for achieving accurate MRE measurements across different anatomical sites in clinical practice.

Significant limitations of the present study should be noted. First, in the phantom experiments, the direction of wave propagation was controlled to occur along the long axis of the phantom, and MRE data were acquired with the phase-encoding direction either parallel or perpendicular to this direction. However, factors such as tissue heterogeneity, organ boundaries, and actuator geometry lead to much more complex patterns of wave propagation in various organs *in vivo*. Second, the *in vivo* liver MRE experiments included only three healthy volunteers, limiting the generalizability of findings regarding distortion effects in clinical populations. Therefore, further work with larger sample sizes and diverse clinical conditions is required to comprehensively understand how geometric distortion affects MRE measurements across various organs and patient populations.

## 5. Conclusion

We investigated the influences of geometric distortion on shear stiffness measurements in MRE and identified two critical factors that compromise measurement uniformity: the spatial uniformity of shear stiffness measurements decreases when the wave propagation direction is parallel to the phase-encoding direction, and when shorter propagating wavelengths occur at higher vibration frequencies or in softer tissue regions. As MRE applications are currently being extended to various organs, we suggest careful attention to these two factors to ensure accurate and reliable MRE.



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## Conflicts of Interest

The authors declare no conflicts of interest regarding the publication of this paper.

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