# Motion Compensation and Super-Resolution in Magnetic Resonance Elastography

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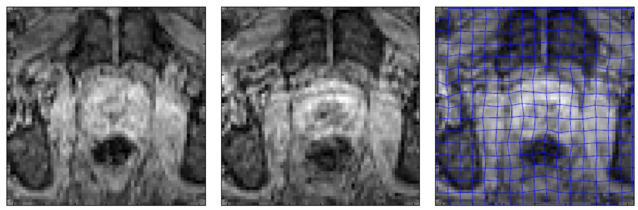
## Introduction

Magnetic resonance elastography (MRE) uses a sinusoidal mechanical excitation and a synchronized pulse sequence that encodes the resulting wave motion of tissue. In conventional MRE[1], each spatial component of the three-dimensional (3-D) displacements is encoded separately, and the wave is sampled at different stages during its cycle by varying the phase shifts between the mechanical excitation and the motion encoding gradients (MEGs) in each acquisition. Although the Nyquist rate requires only two samples per direction to estimate the amplitude and phase of the displacements at each voxel[2,3], typically more samples are acquired to improve signal-to-noise ratio (SNR). Due to oversampling of the wave, an MRE scan with a typical field of view is a lengthy process that may cause inconvenience to the patient, and increases the possibility of both voluntary and involuntary movements. Patient's movements in MRE introduce artifacts, and corrupt displacement estimation and elastogram reconstruction. Thus, there is a trade-off in acquiring more samples to improve SNR, or fewer samples to reduce patient's movements. A compromise is to allow reasonable scanning times by acquiring four or eight samples of lower spatial resolution images. In this work, we utilize the redundancy of images acquired in MRE, and apply image registration to compensate for patient's motion during the scan. We then employ a super-resolution technique to enhance the resolution of the acquired images to allow for a better estimation of tissue displacements and elasticity.

## Methods

In MRE, the phase component of the complex imaging signal, known as the phase image, encodes the displacement, while its magnitude component, known as the magnitude image, is still available as a low-resolution T2-weighted image that is inherently registered to its phase image. Ideally, all magnitude images, acquired in different directions and wave samples, are identical. In practice, they vary due to noise and patient's movements. We propose to compensate motion among different acquisitions of phase images by registering their corresponding magnitude images. We consider an arbitrary magnitude image as the reference image, and register the rest of the $(3K-1)$ magnitude images to it, where $K$ is the number of samples in each direction. We employ a deformable registration algorithm with a sum-of-square differences similarity metric and elastic regularization[4]. We then apply the resulting deformation maps to the corresponding phase images. Next, assuming a uniform sampling of the wave in MRE acquisition, we consider $K/2$ pairs of motion compensated phase images. Each pair, corresponding to acquisitions $k$ and $(k+K/2)$, represents the same wave sample (up to a sign flip and noise). Rather than basic interpolation methods, which introduce artifacts such as blur and contrast loss, super-resolution techniques efficiently increase the native size of images by incorporating information from different images of the same scene[5]. Thus, to each pair, we apply an iterative back-projection super-resolution method with a steepest-descent choice of the back-projection kernel[6]. We then have $K/2$ super-resolution phase images, from which we estimate a super-resolution displacement map and reconstruct the super-resolution elastogram using local frequency estimation[1].

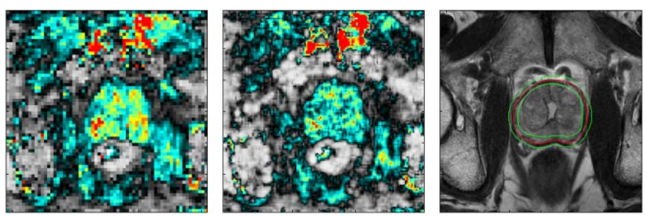


(a) Reference (b) Template (c) Registered template

**Figure 1.** Motion compensation using registration. Cross-section of MRE magnitude volumes from the mid-gland region of a patient's prostate scan. Blue grid represents the deformation field.

## Results

We evaluate the proposed approach on both an elasticity phantom (Model 049, CIRS, VA, USA), and 12 patients' prostate scans (age 65±6 years old). The MRE scans were acquired using a using a gradient echo sequence, and a custom-made shielded electromagnetic transperineal transducer[7] synchronized to a scanner (Achieva 3.0T, Philips, The Netherlands), with 200Hz and 70Hz vibrations applied to the phantom and patients, respectively. We have oversampled the scan of each acquisition with $K=8$ phase shifts of the MEGs ranging evenly between $[0,2\pi]$ on a 128x128x24 grid with an isotropic voxel size of 2mm$^3$. The MRE scan lasted 8 min for a 3-D displacement field acquisition. Since the phantom is static, the proposed super-resolution method was applied to its data without motion compensation. The root-mean-square errors of the factory elasticity values versus the native resolution and the super-resolution elasticity values were 6.6±5.5kPa and 4.7±4.3kPa, respectively. For the patient's data, the normalized cross-correlation among magnitude images was improved by 2.1±0.5% after motion compensation. The contrast and edge strength along the prostate contour in the super-resolution elastogram were, respectively, 1.8±0.8% and 22±4% higher than in the native resolution elastogram.



(a) Native resolution (b) Super-resolution (c) T2-Weighted

**Figure 2.** Super-resolution in prostate MRE. T2 segmented prostate (red contour), and region (green contours) that were used for edge and contrast evaluation. Intensity scale of elastograms is 0–30kPa.

## Discussion

We outlined an image processing method for enhancing MRE data without altering the current acquisition protocol. The approach utilizes unique properties of MRE, namely, the similarity of the "by-product" magnitude images for compensating motion in the phase images, and the redundancy of the phase images for super-resolution. In MRI, super-resolution has been proven to be of limited benefit in-plane due to the bandlimited nature of the Fourier encoded images[8]. However, in our approach, we employ super-resolution on the phase component of the signal, and assume that the nonlinear residual physical movements of the object allow for higher frequency components to be recovered. Phantom experiments and clinical data show improvement in elasticity reconstruction and gland-background separability that may facilitate segmentation and classification of cancer. In some cases, however, there are image artifacts, typically around the base and apex, that might be caused by registration errors. Future work may employ modeling the blurring kernel for super-resolution using the scanner's point spread function rather than a simple Gaussian.

## References

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