Comparison of MR Elastography of the Brain at 1.5 T and 3.0 T

M. C. Murphy[1], K. J. Glaser[1], J. P. Felmlee[1], and R. L. Ehman[1]
[1]Mayo Clinic College of Medicine, Rochester, MN, United States

Introduction

Magnetic resonance elastography (MRE) is a method for quantitatively measuring the stiffness of tissues by following the propagation of shear waves using a phase contrast MRI technique [1]. Several groups are evaluating MRE in the brain as a means for potentially evaluating diffuse disease and focal disease effects through changes in shear stiffness [2-4]. One of the key challenges that has emerged is the high attenuation of shear waves in brain tissue, affecting the ability to observe wave propagation in deep structures. In this context, performing MRE at higher field strengths may offer advantages compared with imaging on a 1.5 T system. The purpose of this work was to directly compare MRE wave imaging results that can be obtained in optimized protocols at 1.5 T and 3.0 T in studies of phantoms and human volunteers.

Methods

MRE phase difference images were collected in a phantom and in normal human volunteers at 1.5 and 3.0 Tesla on GE EXCITE scanners. The phantom was a uniform cylindrical 15% bovine gel, and the *in vivo* data were acquired in a healthy volunteer after obtaining informed consent. In both experiments, shear waves at 60 Hz were introduced with a passive pneumatic driver placed posterior and lateral to the object and connected by a 25-ft plastic waveguide to an active voice-coil system. The location of the driver rocks the phantom or volunteer's head in a 'no' motion. In the phantom studies, the driver and adjacent support were attached to the phantom in order to minimize experimental error intrinsic in moving the phantom between scanners. A single-channel quadrature birdcage coil was used in each case and the automatic prescan routine was used estimate the optimal center frequency and transmit/receive gains for the system. Four phase offsets were collected using a multislice gradient echo pulse sequence. Imaging parameters in the phantom included: TR/TE = 266/26.7 ms, FOV = 24 cm, 8 slices in 1 pass, slice thickness = 2 mm, 1.0 G/cm motion encoding gradient, 120x120 acquisition matrix, 16 kHz bandwidth. Imaging parameters in the volunteer were: TR/TE = 200/26.3 ms, FOV = 24 cm, 8 slices in 1 pass, slice thickness = 3 mm, 2.0 G/cm motion encoding gradient, 128x96 acquisition matrix, 16 kHz bandwidth. Motion was encoded with one motion-sensitizing gradient pair in the anterior-posterior direction. The amplitude of the first temporal harmonic of the wave data was used as a measure of the amplitude of cyclic phase change throughout the object, which is proportional to the displacement amplitude in the object [1]. The same slice was chosen to analyze the phase SNR at the two field strengths. The MR magnitude SNR, which is inversely proportional to the standard deviation of the MR phase signal [5], was calculated as the ratio of the MR signal to the standard deviation of the noise. The phase SNR was calculated as the ratio between the amplitude of the first harmonic of the wave data and the standard deviation of the phase signal. The mean and standard deviation of the phase SNR data in the phantom were calculated from regions of interest that included the entire central slice. Differences in phase SNR were tested by ANOVA.

Results

Example phase difference images in the phantom along with phase SNR maps can be seen in Figure 1. The phase SNR in the phantom at 1.5 T was 178.3 ± 78.1 (mean ± standard deviation), while at 3.0 T it was 291.0 ± 132.1. Phase SNR was significantly different between the 1.5 and 3.0 T cases as assessed by ANOVA ($p<0.01$). Example phase difference images in the volunteer and their magnitude SNR maps are shown in Figure 2.

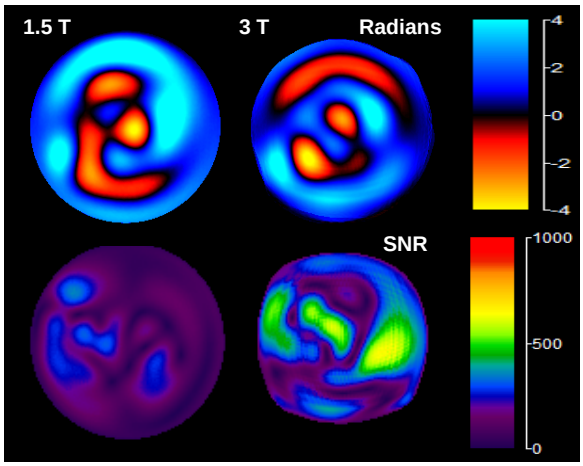

Figure 1. Example phase difference images (top) and phase SNR maps (bottom) in a phantom.

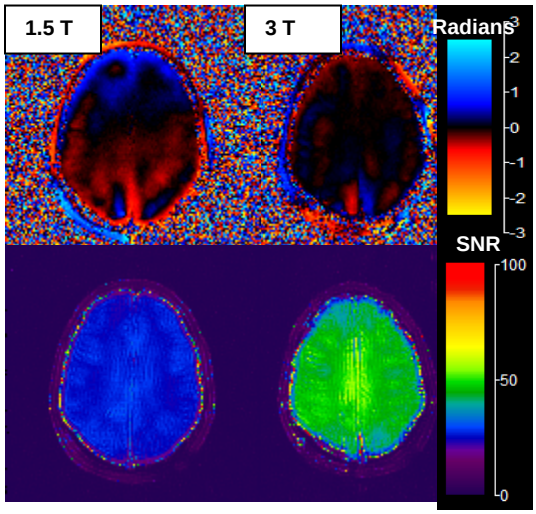

Figure 2. Example phase difference images (top) and magnitude SNR maps (bottom) *in vivo*.

Discussion

Despite the fact that MRE data analysis is based on only the phase of the measured MR magnetization, field strength is important to phase SNR and thus data quality. To understand this phenomenon, consider measuring complex transverse magnetization with some constant uncertainty in the real and imaginary parts of the signal. As the magnitude of the magnetization increases, the range of possible measured phases decreases. This effect is demonstrated in Figure 3. Since magnitude SNR increases with field strength in the brain (Figure 2), we also hypothesize that phase SNR will be improved in the brain by collecting data at 3.0 T. This gain in SNR is important to the optimization of MRE in the brain as it can improve the reproducibility of measurements or be traded through various strategies for shorter acquisition times or increased spatial resolution.

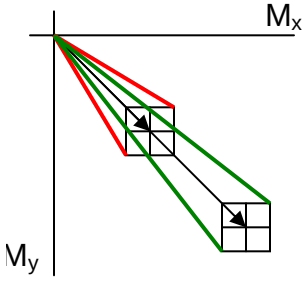

Figure 3. Demonstration of the effect of increased magnetization magnitude on phase error.

References

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