MRE of In Vivo Human Lung Parenchyma: Feasibility Study of Motion Encoding using the Imaging Gradients with ^{1}H MRI

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Introduction: It is appreciated that lung form and function are the result of the dynamic interplay between the intrinsic mechanical properties of lung parenchyma, surface tension and transpulmonary pressure and that disease processes play a fundamental role in perturbation of this balance [1] Magnetic Resonance Elastography (MRE) is a phase-contrast MRI-based elasticity imaging technique that can assess the mechanical properties of soft tissues by synchronizing motion encoding gradient waveforms with an external source of shear wave excitation [2]. The most practical method of implementing MRE within the lung is by modification of existing ^{1}H MR imaging techniques [3]. However, this is technically challenging due to the ultra-short T2* of the lung (~ 1- 2 ms). A significant challenge in ^{1}H lung MRE is the increased echo time (TE) of the imaging sequence due to the requirement of the additional motion-sensitizing gradients (MSG), resulting in poor tissue signal. It was hypothesized that the existing imaging gradient waveforms can be modified to have sufficient shear wave motion sensitivity while maintaining a low echo time. The purpose of this work was to test this hypothesis by imaging shear wave motion in the lungs of a normal human volunteer with a modified spin echo based MRE imaging sequence.

Methods: All experiments were conducted according to institutional review board guidelines using a 1.5-T whole-body scanner (Signa Excite, GE Healthcare, Milwaukee, WI). A pressure-activated passive drum driver, connected to an active acoustic speaker, was placed above the right lung on the anterior body wall of the volunteer, the position of which is indicated schematically in Fig. 2a. 60-Hz continuous vibrations were introduced into the lungs. Sagittal images of the right lung away from the heart were obtained to reduce the effects of the cardiac motion. Images were acquired with a modified spin echo MRE pulse sequence (Fig. 1). The major modification of this sequence are the two crusher gradient lobes (arrows) which function together as a motion-encoding gradient pair whose sensitivity is affected by their duration and their separation in addition to spoiling residual transverse magnetization. Design of these lobes is determined by the interplay between increased motion sensitivity achieved by increasing gradient lobe duration versus minimizing TE which increases with increasing gradient waveform duration. Another important parameter is the separation of these two lobes in comparison to the period of the mechanical motion. The crusher lobes are minimally sensitive to the motion if their separation is an integer (k) multiple of the period of motion, are maximally sensitive to the motion if it is (k + ½) multiples of the period of motion, and there is a gradual sensitivity variation in between. For this study, the crusher gradient lobes were 2.5 ms long and separated by 5 ms (0.3 * period of 60 Hz motion). Other imaging parameters were FOV = 34 cm, acquisition matrix = 128x64, frequency-encoding direction = SI, motion sensitization direction = RL, TR/TE = 400/8 ms, slice thickness = 10 mm, 3 phase offsets, acquisition time = 3 breath holds of 21 sec.

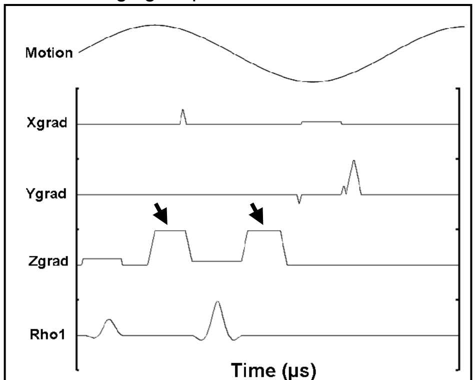

Figure 1: The spin echo Pulse sequence. The crusher gradients (indicated by the arrows) are used for the motion encoding.

Results: Figure 2 shows the results obtained from the volunteer study. Figure 2a shows the magnitude image of the sagittal slice of the right lung with the driver position and direction of motion indicated. Figure 2b shows a single phase offset image of the same slice without any applied motion. Coherent and homogenous phase within the lung can be seen. Additional, background phase is also observed in the absence of motion. In conventional MRE, this background phase is removed by acquiring the same image with an inverted polarity of the MSG and taking a phase difference (similar to phase-contrast imaging). In this study, an acquisition was not performed with the polarity of the crusher gradients inverted, thus the background phase is still present. The phase image of the same slice with the motion on is shown in Fig. 2c. Compared to Fig. 2b, the additional phase due to the propagating shear waves is visible. The background phase can be removed either by Fourier transformation through the temporal phase offsets since the background phase is constant or by subtracting off the phase data obtained without the applied motion. Figures 2d shows the real part of the fundamental harmonic after the removal of the background phase (shown in displacement amplitude in microns). While the data are noisy, the shear wave propagation within the lung (see anterior portion indicated by the arrow) can be easily visualized from this image. Density weighted shear modulus (μ/ρ) map within the lung where shear waves are visible was calculated to be 2.7 ± 0.73 kPa using the local frequency estimation algorithm [4]. In this study only motion occurring in the through-plane direction (RL for a sagittal acquisition) was measured. However, it is possible to replicate the crusher gradients on the other 2 axes to allow for motion encoding of the other directions without increasing the TE and to enable phase difference calculations.

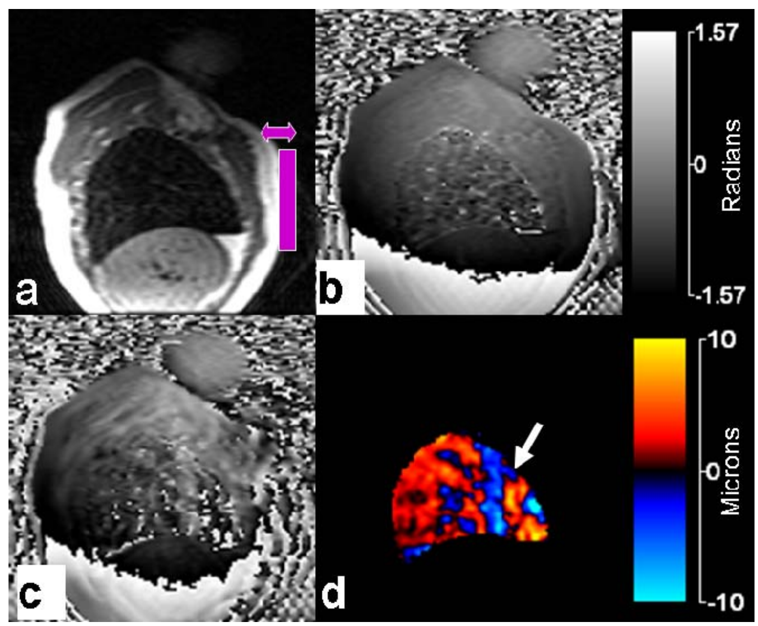

Figure 2: (a) Sagittal magnitude image of the right lung, the passive driver position and the vibration direction are indicated schematically. (b) Phase image of the same slice without motion, coherent phase within the lung is visible. (c) Phase image with motion, shear-wave-dependent phase is visible. (d) Real part of the first harmonic image of the three phase offsets of the segmented lungs (in displacement amplitude in microns), the shear waves propagating from the anterior lungs is easily visible (arrow).

Conclusion: These data suggest that it is feasible to use the imaging gradients of a spin echo sequence to encode shear motion within the lungs in vivo. Further work is required to increase the flexibility of the acquisition and to investigate the clinical potential of MRE of the lungs.

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References: (1) Suki et al., J Appl Physiol 98(5) : 1892-1899, 2005. (2) Muthupillai et al., *Science* 269: 1854-1857, 1995. (3) McGee et al., JMRI 29(4): 838-45, 2009. (4) Manduca A et al., *Med Imag Anal*. 5: 237-254, 2001