TREMR: Table-Resonance Elastography with MR

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Introduction: MR Elastography (MRE) has been demonstrated to offer the capability of measuring tissue stiffness non-invasively, with a wide variety of potential applications [1]. Tissue stiffness is inferred from motion-encoded MR imaging of mechanical vibrational waves as they propagate through the tissue. A factor hindering widespread implementation of MRE is that the majority of methods in the literature require the use of specialist hardware to induce the mechanical vibrations. Previous investigators performing MRE of the human brain have developed elaborate hardware which transfers the vibration to the head via a bite-bar [1, 2, 3]. However, these devices tend to be uncomfortable and not all subjects tolerate bite-bars [4]. Alternatively, the large gradient lobes employed in diffusion-weighted imaging have been shown to cause low-frequency vibrations (20-25 Hz) of a clinical MR system, including the patient table, of amplitudes on the order of ~100 µm [5]. As this a similar magnitude to the vibrations typically used for MRE, we sought to determine if vibrations induced in the patient table via gradient-switching could be used as a mechanical driver for MRE.

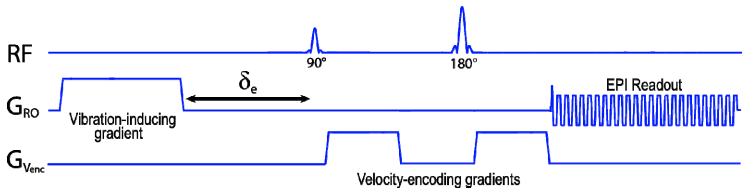

Fig. 1: *Pulse-sequence diagram for proposed MR Elastography implementation.*

Implementation: Figure 1 shows the proposed pulse-sequence for MRE which uses a 35 mT/m gradient applied along the left-right axis with a duration of 20 ms to excite mechanical vibrations close to 25 Hz. There is then a variable delay, which we refer to as δ_e, before a spin echo sequence with velocity-encoding gradients which impart phase to the image proportional to the local velocity in the direction of the gradients. An EPI readout follows velocity encoding. Imaging was performed on a Siemens 3T TIM Trio system with a single-channel head coil. EPI readout parameters: 64x64 matrix, 3 mm isotropic resolution, TE = 56 ms, v_{enc} = 1.5 mm/s. To allow the vibrations from the previous acquisition to decay sufficiently the TR was double the shortest possible, giving a minimum TR per slice of 150 ms. The vibration-inducing gradient was always applied along the left-right direction as we have found this to cause the strongest vibrations of the patient table. The velocity-encoding gradients, however, were applied separately along all three axes to allow measurement of the full 3D velocity vector. δ_e was varied between 0 and 72 ms, also extending the TR per slice by the same amount. To ensure good transfer of mechanical vibrations from the patient table to the head, padding was inserted between the headphones and the coil in the same manner as is normally used to restrain head movement.
Images were reconstructed offline using Matlab (The MathWorks) and phase unwrapped using PRELUDE software [6].

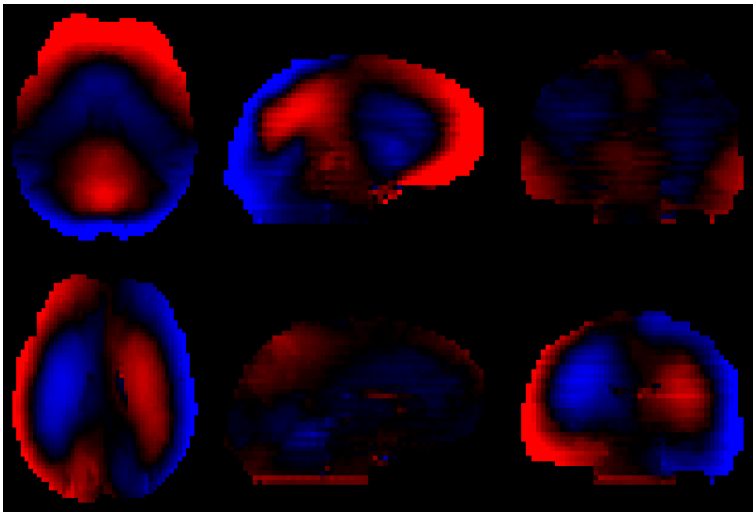

Fig. 2: *Axial, sagittal and coronal sections of the unwrapped phase at δ_e = 24 ms with velocity-encoding in the left-right direction (top), and the anterior-posterior direction (bottom).*

Results: Figure 2 shows sections of the unwrapped phase for a single delay time (δ_e = 24 ms) with velocity encoding in the left-right and anterior-posterior directions for a healthy 25 year-old male subject. Clear motion was also measured in the superior-inferior direction, but of lower magnitude (not shown). Peak measured velocities were ~6 mm/s. By integration of velocities the peak displacement was found to be ~90 µm.
Figure 3 shows the 2D velocity vector field from a single slice as δ_e is increased from 0 to 72 ms, depicting wave propagation. It is clear from these vectors that the table vibrations have imparted rotational movement to the skull. This movement is transmitted as a shear wave through the brain tissue. A similar wave-pattern was observed in multiple subjects and is comparable to that observed by others using a bite-bar apparatus to provide the vibrations (cf. [1,2,3]).
Discussion and Conclusion: We have successfully demonstrated that the inherent vibrations of a clinical MR system in response to low-frequency gradient switching can be transmitted into the brain with sufficient magnitude to image the wave propagation in 3 dimensions. We have yet to determine if accurate stiffness maps may be derived from these data, however, as the driving function is not at a single known frequency as is usually imposed when performing MRE with a separate mechanical driver. Measurement of the table vibrations using laser interferometry, such as in [5] for diffusion imaging, may help with these calculations. A further restriction of the technique is that the vibrations can only be generated at mechanical resonant frequencies of the system. Although in this preliminary experiment we chose to image the brain, the same approach could be employed for other areas of the body, provided there is sufficient mechanical coupling between the patient table and the tissue of interest.

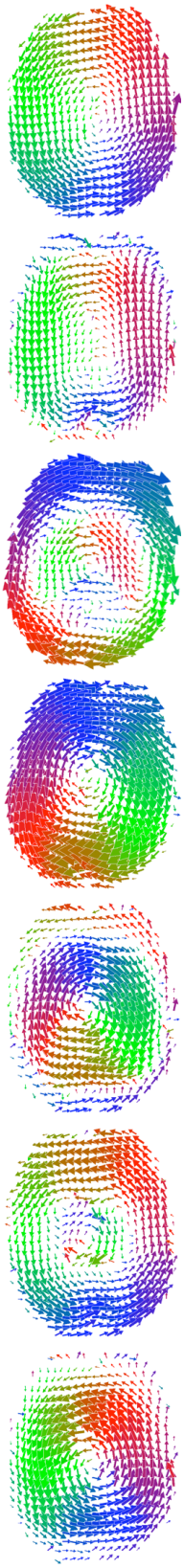

Fig. 3*: 2D velocity vectors as δ_e was varied from 0 ms (top) to 72 ms (bottom) in a single slice. Colours encode direction of vector; size of arrow encodes magnitude of vector. (For animation see http://tinyurl.com/tremr)*

[1] A Manduca *et al*, Med. Im. Anal. 5 (4) 237-54 (2001); [2] L Xu *et al*, Acta Rad. 48 (3) 327-30 (2007); [3] SA Kruse *et al*, NeuroImage 39 (1) 231-7 (2008); [4] P. Jezzard *et al* (eds). FMRI: An Introduction to Methods. OUP (2003). [5] J Hiltunen et al, NeuroImage 32 (1) 93-103 (2006); [6] M Jenkinson, MRM 49 (1) 193-7 (2003)