Magnetic Resonance Acoustic Radiation Force Imaging (MR-ARFI)

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Introduction: Acoustic radiation force imaging is an imaging method that maps the displacements induced by focused ultrasound pulses. By dynamically focusing at multiple locations using a phased array transducer, one can produce images of the displacement that can be related to mechanical tissue properties. Previous work with this modality has used ultrasound imaging (1,2). The purpose of this work was to test the feasibility of magnetic resonance acoustic radiation force imaging (MR-ARFI).

Methods: This method uses quasi-static MR elastography (3) to measure the displacements induced by a focused ultrasound pulse and utilizes a linescan sequence previously developed for diffusion MRI (4). In contrast to previous investigations using 1D MR techniques that measured shear waves emanating from an ultrasound pulse (5,6), we are interested in simply mapping the magnitude the displacement at the focus for a single pulse. A 1.63 MHz focused ultrasound transducer (radius of curvature/diameter: 8/10 cm) supplied the pulses. During each line acquisition a single 42 ms pulse was delivered 6 ms after the start of the 90° RF slice select pulse (Fig. 1). The pulse started approximately 4 ms before the motion encoding gradient and ended immediately after the 180° pulse. Linescan parameters were: voxel dimensions: 0.8×2×3 mm; TE: 86.6 ms; encoding gradient strength (G_e), duration (Δt): 28 mT/m, 22 ms; field of view: 20 cm; number of frequency encodes: 256, signal averages: 5. Multiple (10-20) lines were acquired to maps of the displacement. The time between line acquisitions was 400 ms. Displacements were estimated by multiplying the phase difference between two acquisitions with inverted motion encoding gradients by $\gamma G_e \Delta t$. With the parameters as implemented, the sensitivity of the sequence to displacement was thus 6.0 µm/radian. A 1.5T clinical MRI and an 18 cm diameter receive-only surface coil were used. The feasibility of detecting the focal displacements was examined in a homogeneous silicon tissue-mimicking phantom (GE RTV6166 mixed in 30/70 ratio, density: 1100 kg/m³, sound speed: 1050 m/s: attenuation: 3.45 Np/m/MHz, Young's modulus: 15.3 kPa) and in ex vivo bovine kidney for a range of acoustic power levels.

Results: Focal displacements were readily observed in both the phantom and in the kidney samples (Fig. 2). Displacements produced by pulses at acoustic power levels as low as 0.4 W could be detected. The displacement increased as a function of the acoustic power (Fig 3). At high acoustic powers, the displacement rapidly increased, apparently resulting from irreversible changes that were seen in the silicon phantom after sonication. Phase artifacts presumably produced by vibrations produced by the rapidly switching encoding gradients could be corrected by subtracting off phase changes in regions outside of the focal zone. The standard deviation of the phantom and kidney without displacement was less than ±0.02 radians, which corresponds to a noise level of better than ±0.1 µm.

Discussion: While further work is necessary to validate the measurements, these experiments demonstrate the feasibility of detecting small displacements induced by low power short ultrasound pulses. Using a linescan acquisition will allow for the displacement at a location to be detected with a single ultrasound pulse, in contrast to imaging with phase encoding that would require a pulse for each phase encode. Thus, one could scan the focal coordinate in a raster pattern and rapidly produce displacement maps while maintaining a safe ultrasound exposure. MRI guided focused ultrasound systems that utilize phased array transducers are becoming a clinical reality (7) and could be used as ultrasound sources for such imaging. The ability to detect and quantify short ultrasound pulses may also be useful for monitoring ultrasound therapies that use low acoustic intensities and/or pulsed exposures, such as targeted drug delivery, stroke therapy, and cavitation-based ablation therapies. Limitations of the method as implemented include its long TE and vibration artifacts induced by the rapid gradient switching. These limitations might be reduced by optimizing the sequence by using shorter, non-rectangular gradient trajectories.

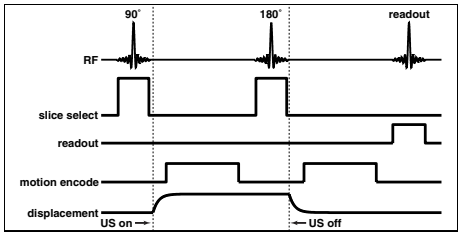

Fig 1: Pulse sequence diagram and assumed focal displacement as a function of time induced by the ultrasound pulse.

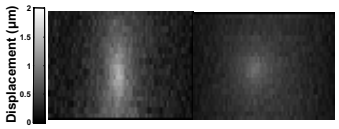

Fig 2: Maps showing the focal displacement in the US beam direction from a 42 ms ultrasound pulse in a phantom. Left: map along direction of ultrasound beam, 5W; Right: map in the focal plane, 2.5W.

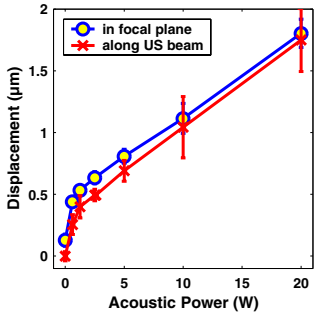

Fig. 3: Estimated displacement as a function of acoustic power in the phantom. At 40W, the displacement jumped to 5.4-12.4 µm, presumably due to irreversible changes induced in the phantom that were observed after the experiment.

References

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