Experimental Validation of a 3D MR-Acoustic Radiation Force Imaging Simulation Algorithm

Allison Payne[1], Joshua de Bever[2], Alexis Farrer[3], Dennis Parker[4], and Douglas Christensen[3]
[1]Radiology, University of Utah, Salt Lake City, UT, United States, [2]Computer Science, University of Utah, Utah, United States, [3]Bioengineering, University of Utah, Utah, United States, [4]Radiology, University of Utah, Utah, United States

PURPOSE

In magnetic resonance guided focused ultrasound (MRgFUS) therapies, the *in situ* characterization of the focal spot location and quality is critical. MR acoustic-radiation-force-imaging (MR-ARFI) is a technique that measures the tissue displacement caused by the concentrated force exerted by the FUS beam. During the pre-treatment phase, MR-ARFI can assess the focus and targeting quality of the FUS beam as well as be used to correct for phase aberrations. This work presents the experimental validation of a 3D simulation method that models both the radiation force pattern and resulting displacement field. *TARGET AUDIENCE:* Physicists and clinicians wanting a more quantitative measure of beam patters for use in MRgFUS.

$$F = \frac{2\alpha I}{c}\Delta x \Delta y \Delta z + \frac{2IR^2}{c}\Delta x \Delta y \quad \textbf{(1)}$$

$$R = (\rho_2 c_2 - \rho_1 c_1)/(\rho_1 c_1 + \rho_2 c_2) \quad \textbf{(2)}$$

$$I = \left|p^2\right|/2\rho c \quad \textbf{(3)}$$

$$w = \frac{F}{8\pi\mu}\left(\frac{z^2}{r^3} + \frac{1}{r}\right) = \frac{F}{8\pi\mu} g(r) \quad \textbf{(4)}$$

METHODS: *Simulation Theory*

The force distribution due to the radiation force of the ultrasound beam is found by summing the forces due to absorption in each voxel and due to reflection at the interface between two voxels (Eqs. 1 & 2). Each voxel has an assigned absorption coefficient (α, Np/cm), density (ρ, kg/m^3) and speed of sound (c, m/s). The beam intensity pattern I (Eq. 3), is found using the Hybrid Angular Spectrum method,[1] which rapidly propagates the pressure pattern along the beam axis by alternating between the space and spatial-frequency domains. When a steady-state point source force acts internally in an infinite homogeneous medium, the displacement of the material in all directions is given by the Somigliana elastostatic tensor[2]. Assuming the displacement is predominantly in the direction of the ultrasound beam and assuming linear isotropic properties, an incompressible medium and steady-state conditions, the displacement w in the direction of beam propagation reduces to Eq. 4; a convolution of the force field pattern from Eq. 1 and a 3D Green's function $g(r)$, where μ is the Lamé shear constant ($\mu = E/3$), r is the distance from the point source to the location of w and z is the projection of r onto the z-axis. This convolution operation, denoted by the * symbol in eq. 4, is accomplished in the spatial-frequency domain. While the theory assumes that steady-state conditions exist, the phase accumulates dynamically during the encoding duration of the MR-ARFI sequence; in order to account for this, a linear filter was applied to $g(r)$ before convolution.

Phantom stiffness, bloom	Young's modulus E, kPa (mean ± σ)	Attenuation, Np/cm (mean ± σ)
125	9.5 ± 1.8	0.31 ± 0.05
175	18.8 ± 2.7	0.31 ± 0.08
250	29.4 ± 4.7	0.36 ± 0.05
ρ = 1000 kg/m^3, c = 1550 m/s		

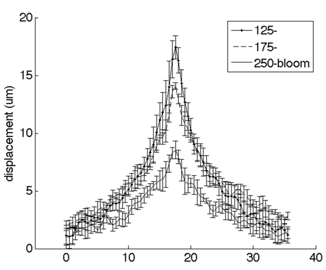

***Figure 1**: Experimental MR-ARFI in plane displacement (errorbars=1 s.d.)*

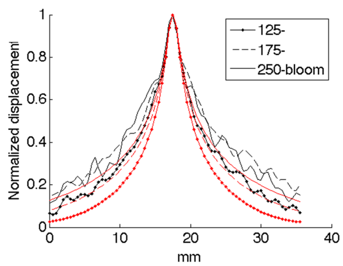

***Figure 2**: Comparison of normalized simulated (red) and experimental (black) displacement patterns for all phantoms*

Experimental Protocol: Experimental validation used gelatin-based phantoms of varying mechanical stiffness (125-, 175- and 250-bloom). The gelatin mixtures were made with a concentration of 50% evaporated milk in order to achieve an attenuation coefficient similar to that expected in tissue. The Young's modulus and attenuation values for each phantom type were obtained using independent measurement techniques (Table 1). All experiments were performed with a pre-clinical MRgFUS system (IGT, Inc., Pessac, France) with a 256-element phased-array transducer (f = 1 MHz) in a Siemens Trio 3T scanner. Displacements in the phantoms were measured using a 3D spin echo segmented-EPI sequence[3] with unbalanced-bipolar motion encoding gradients and flyback readout (TR = 250 ms, TE = 50 ms, fat saturation, FA = 90°, EPI = 7, 256x128x36-mm FOV, 2x2x3-mm ZFI'd to 0.5-mm^3 spacing, motion encoding amplitude (ME_{amp}) = 30 mT/m, US_{dur} = 10 ms, US_{power} = 60 W, acquisition time=64 s, N_{ave} = 4).

RESULTS

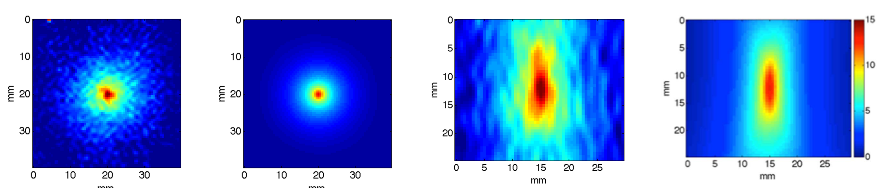

***Figure 3**: Coronal and axial views of the experimental (1st & 3rd frames) and simulated (2nd and 4th frames) displacements in the 125-bloom phantom. Colorbar in μm.*

The means of the experimentally measured MR-ARFI displacement patterns (N=5) in the transverse plane are shown in Figure 1. The normalized experimental displacements are shown along with the corresponding simulated patters in Figure 2. Slice views of the displacement in both the transverse and slice directions for the 125-bloom phantom are shown in Figure 3.

DISCUSSION & CONCLUSIONS

The 3D MR-ARFI sequence used in this work allows for a thorough evaluation of the presented simulation theory. While the general trends and shape of the ARFI displacement patterns agree between experiment and simulation, there are discrepancies in the peak displacement (2.7-6.5 um error). In addition, the simulated displacement pattern has a steeper descent than what is observed experimentally. This may be due to discrepancies in the property measurements or the assumption of homogeneous, linear isotropy. Despite these discrepancies, these promising results demonstrate the potential to use quantitative MR-ARFI displacement data in MRgFUS therapies.

REFERENCES

1. Vyas, U. et al., IEEE Trans Ultrason Ferro Freq Cntrl 2012; 59(6):1093.
2. Aki and Richards, Quantitative Seismology, 2009.
3. De Bever, J. et al., ISMRM 2012, #2921.

ACKNOWLEDGEMENTS

This work was funded by NIH 1R01EB013433, 1R01CA172787, 1R01CA134599, the Focused Ultrasound Surgery Foundation and the Margolis Foundation.