MR Acoustic Radiation Force Imaging: Comparison of Encoding Gradients

J. Chen[1,2], R. Watkins[2], and K. Butts Pauly[2]

[1]Dept. of Electrical Engineering, Stanford University, Stanford, CA, United States, [2]Dept. of Radiology, Stanford University, Stanford, CA, United States

Introduction MR-guided high intensity focused ultrasound (HIFU) is promising for a variety of therapeutic applications including tumor ablation and targeted drug delivery (1). For tumor ablation, HIFU is used in "thermal mode" (continuous wave) and guided by proton resonance frequency (PRF) thermometry, which measures tissue temperature and allows a calculation of thermal dose. For targeted drug delivery, HIFU is used in "mechanical mode" with higher power pulses and a low duty cycle. This allows for a mechanical interaction with tissue, the blood brain barrier, or circulating liposomes, without significant tissue heating. Instead of imaging tissue temperature, MR monitoring may be done by measuring displacements caused by the acoustic radiation force. When the ultrasound is applied during motion sensitizing gradients, such as one of a pair of Stejskal Tanner gradients, the displacement is encoded into the phase of the image. Since the displacement is small (on the order of microns), it is necessary to use either averaging (2) or large motion sensitizing gradients (3). Unfortunately, large Stejskal Tanner gradients may render the image susceptible to eddy currents and signal loss from diffusion. The purpose of this work was to compare several gradient configurations for MR acoustic radiation force imaging (MR-ARFI).

Methods Imaging was performed with a line scan sequence (TR/TE=500/69 ms, FOV=24×6 cm, matrix size=256×63, slice thickness=5 mm, bandwidth=7.81 kHz, nex=5) on a 3T GE Signa MR scanner equipped with an MR-compatible HIFU system (InSightec, Israel). Three gradient sets were compared as shown in Figure 1: (a) two single lobes, (b) repeated bipolars and (c) inverted bipolars. The maximum gradient amplitude (4G/cm) was used for all three sets, and the total effective motion encoding time was kept to 23.6 ms.

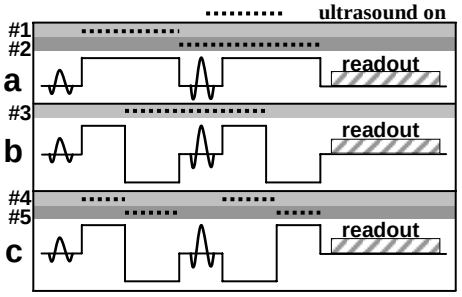

Fig.1 Three motion encoding gradient sets for comparison (a-c), and sonication schemes (#1-#5) used with different gradient sets. The total effective encoding time was 23.6 ms for all methods.

The HIFU system was triggered by the MR sequence to emit ultrasound pulses (80W electrical power) synchronized with the encoding gradients, the timing of which is given by dotted lines in Figure 1. For gradient set (a) and (c), two different sonication schemes were applied: sound on either the 1st or the 2nd lobe/halves. A total of five different acquisitions (#1-#5) were compared. The 1.0 MHz HIFU transducer array was focused at a depth of 12 cm in a gel phantom, with an ultrasound duty cycle of less than 5%. To obtain the background phase map, images were acquired without the application of the ultrasound.

Pairs of images were scanned, with identical sonication timing but opposite polarity of the position encoding gradients. For each pair, the phase difference was calculated and then corrected by a two-step algorithm. In step one, a linear regression was performed outside the focal spot for each line along the readout direction, then the resulting constant and linear phase terms were removed from the whole line. The subtracted linear and constant terms changed from line to line, probably induced by bulk motion from small vibrations during the acquisition. In step two, the 3 background phase maps of gradient sets (a), (b) and (c) were examined and subtracted from the corresponding step one output. After the two steps, the residual phase is proportional to the acoustic displacement with sensitivity of 1.98 μm/radian using the imaging parameters above.

Results All phase images are illustrated with same display range ([-0.1 0.1] radians). The background phase map is shown in Figure 2. From this image, it is clear that the repeated bipolar encoding gradients (scheme b) significantly reduce the nonlinear background phase. In addition, the b value is greatly decreased from ~1300 s/mm^2 in scheme (a), to ~400 s/mm^2 in scheme (b) and ~250 s/mm^2 in scheme (c). Therefore the signal noise ratio (SNR) is enhanced at no cost to displacement sensitivity, as demonstrated in Fig.3 (a). To quantify the SNR, a large region of interest (ROI) was chosen outside the focal spot and the standard deviation of the phase (σ_n) was measured, which significantly decreased from ~0.018 radians for gradient set (a) to ~0.009 radians for gradient sets (b) and (c). At the focal spot, an oval-shape ROI was chosen with a size of 40 pixels to measure the mean displacement. The result is plotted in Fig.3 (b), with the displacements ranging from 0.14 μm to 0. 20 μm depending on the sonication scheme.

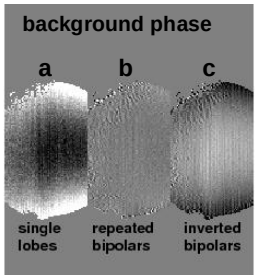

Fig.2 Background phase maps from gradient sets (a)-(c). The background phase is minimized with repeated bioplars, which alleviates the need for additional phase correction.

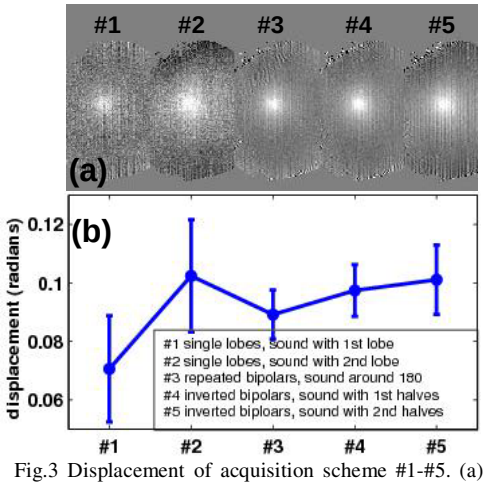

Fig.3 Displacement of acquisition scheme #1-#5. (a) displacement maps after the two-step phase correction, (b) measurements at the focal spot. The SNR is significantly improved by the bipolar gradients.

Discussion This study demonstrates the benefits of applying repeated bipolar gradients to encode displacement in MR-ARFI. The SNR is significantly enhanced at no cost of scan time or displacement sensitivity. A nonlinear background phase correction is not required, as was done in previous work (4). Variations in the measured displacements, despite the same encoding times, could be due to a delay in the tissue response. For example, scheme #2 provides more time for the tissue to respond before the encoding gradient, compared to scheme #1. Future work will include measurement of this tissue response and appropriate optimization of the sequence.

Acknowledgements The authors would like to thank Nathan McDannold, Ph.D., for helpful discussions, and Yoav Medan, Ph.D., for technical support with the HIFU system. This work was supported by NIH RO1 CA111981, RO1 CA121163 and P41 RR009784.

Reference 1) Mitragotri. Nat Rev Drug Discov, 2005;4(3):255-260
2) Yuan, et al. Phys Med Biol, 2007; 52(19):5909-5919 3) McDannold, et al. 15th ISMRM 2007, #967
4) McDannold, personal communication