Fluctuating Equilibrium MR-ARFI

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Introduction: MR-ARFI (acoustic radiation force imaging) is a promising technique for visualizing the focal spot in MR-guided focused ultrasound (FUS) without actually heating the tissue[1]. The acoustic radiation force displaces the tissue slightly at the ultrasound focus, and this effect can be detected by using displacement-encoding gradients to generate a phase shift in an MR image. To date, MR-ARFI has been accomplished using either spin-echo or spoiled gradient-echo pulse sequences. We present here a new approach for generating sensitivity to the ARFI effect, based on a fluctuating-equilibrium (FE) balanced steady state free-precession pulse sequence[2], that results in a magnitude difference (instead of a phase shift) between two oppositely encoded images.

Methods: Our FEMR-ARFI pulse sequence combines a fluctuating-equilibrium phase-cycling scheme with ARFI displacement-encoding gradients, which are applied every other TR window (Fig. 1). Brief (1~2ms) sonication pulses are synchronized with the displacement gradients, with alternating encoding direction. This arrangement results in unbalanced precession angles at the location of the ultrasound focus, which drives the magnetization into an anti-symmetric steady state that yields a signal difference when the two FEMR images are subtracted. The magnitude of this signal difference increases with ARFI-induced phase angle (Fig. 2).

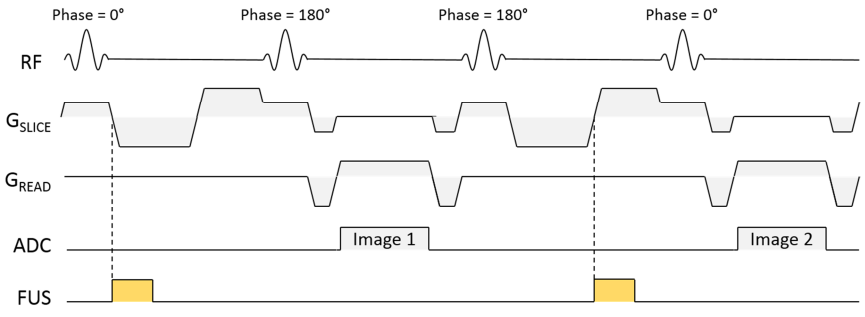

Fig. 1: One cycle of the FEMR-ARFI pulse sequence. Phase-encoding gradients are not shown, to conserve space. FUS pulses are synchronized with the displacement-encoding gradients, with alternating polarity. The same phase-encoding line is acquired twice in each cycle. The cycle is repeated N times to form two N×N images. The ARFI displacement image is reconstructed from the coherent difference between these two (oppositely sensitized) acquisitions.

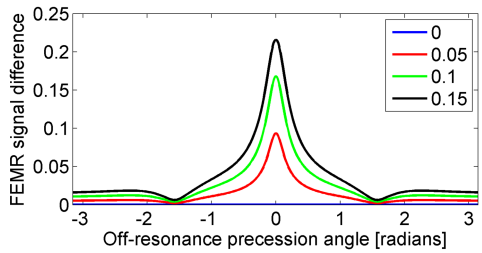

Fig. 2: Results of a Bloch-equation simulation, showing the magnitude of the coherent FEMR signal difference as a function of off-resonance precession angle, for several ARFI-induced phase angles (given in radians). T1 = 900 ms, T2 = 90 ms, TR = 10 ms, and flip angle = 20° were assumed.

To demonstrate this effect experimentally, we performed tests in both a gel phantom and a living rat brain, using an MR-compatible 1 MHz focused ultrasound system with integrated RF coil (RK-100, FUS Instruments Inc., Toronto) and 3T whole-body scanner (Siemens Trio). FEMR-ARFI pulse sequence parameters included: TR = 10ms, TE = 5ms, TH = 3.5 mm, displacement-encoding gradient amplitude = 38 mT/m; phantom: FOV = 96 mm, matrix = 96 × 96; rat: FOV = 70 mm, matrix = 128 × 128. FUS pulse duration was 1 ms (phantom) or 2 ms (rat), with nominal peak-negative pressure of 6 MPa. Following the ARFI measurement in the rat, a standard FUS procedure[3] for opening the blood-brain barrier (BBB) was performed at the same transducer position, to demonstrate effective targeting with the FEMR-ARFI technique.

Results: Figure 3a shows a representative FEMR-ARFI image in the gel phantom. The image magnitude is expected to be zero everywhere except at the location(s) of ARFI-induced tissue displacement. In addition to the clean focal spot at the center, we appear to be picking up the shear wave emanating radially outward from the focal spot as the FUS transducer is pulsed on and off at approximately 50 Hz. Fig 3b shows a high-resolution spoiled gradient-echo, gadolinium-enhanced image of the rat brain before performing the BBB-opening sonication; a superimposed region-of-interest from the FEMR-ARFI image; and a gadolinium-enhanced image acquired after sonication, showing focal BBB opening at the target location.

Conclusions: We have demonstrated both in vitro and in vivo, a new pulse-sequence method for performing MR-ARFI that encodes ARF-induced tissue displacement as a coherent signal difference between two FEMR images with opposite RF phase. Because the underlying steady-state free-precession pulse sequence is inherently fast, this technique may lend itself to rapid 3D MR-ARFI measurements. As shown in Fig. 2, off-resonance conditions suppress the desired effect, so care must be taken to ensure that the magnetic field is on resonance in the region of interest.

References:

[1] McDannold and Maier, Med Phys 35:3748-58 (2008).

[2] Vasanawala et al., Magn Reson Med 42:876-883 (1999).

[3] Timbie et al., J Acoust Soc Am 134:4047 (2013).

Acknowledgments:

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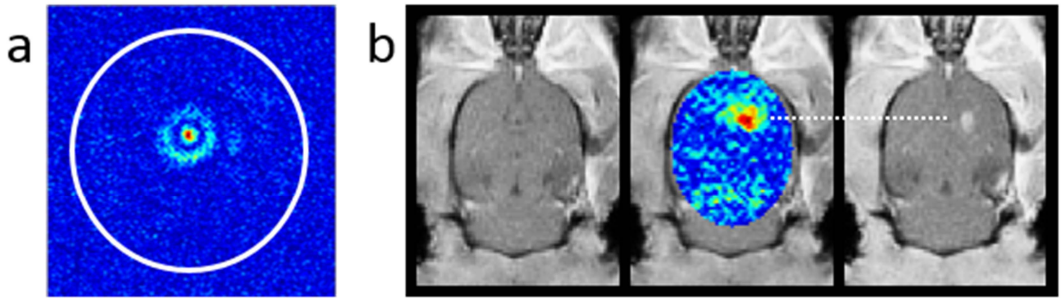

Fig. 3: (a) FEMR-ARFI image in the gel phantom. The boundary of the 3-inch diameter cylindrical phantom is shown for reference. **(b)** Gadolinium-enhanced image of the rat brain before performing the BBB-opening sonication; superimposed region-of-interest from the FEMR-ARFI image; and a gadolinium-enhanced image acquired after the sonication, showing focal BBB opening at the targeted location.