Evaluation of bipolar encoding configurations for spin echo MR-ARFI

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Introduction

MR-based Acoustic Radiation Force (MR-ARFI) has been recently proposed as a tool for the targeting, monitoring and assessment of focused ultrasound (FUS) therapy [1- 6]. Several bipolar encoding schemes have been introduced for spin echo based MR-ARFI [3-5](Fig. 1). The purpose of this work was to evaluate the performance of three encoding schemes by modeling and experimentally testing them in tissue mimicking phantom and *ex vivo* brain tissue.

Background

$$\phi(t) = \gamma \int_{t_{on}}^{t_{off}} G(u) \cdot x(u) \cdot du \quad (1)$$

MR-ARFI measures the displacement of tissue in the focal spot due to the acoustic force of the FUS beam [1]. If tissue displaces as *x(t)*, the simultaneously applied displacement encoding gradient G encodes the tissue displacement as phase $\Phi(t)$ (Eq. 1).

$$x(t) = 0, \quad t \le 0$$
$$x(t) = \frac{F}{k}\left(1 - \exp\left(-\frac{t}{\tau_{rise}}\right)\right), \quad 0 \le t \le T_{off} \quad (2)$$
$$x(t) = x(T_{off}) \cdot \exp\left(-\frac{t}{t_{decay}}\right), \quad t > T_{off}$$

Depending on the viscoelastic properties of tissue, tissue responses to the force according to the model in Eq. 2 [7], where *k* is a proportionality constant relating acoustic force and displacement, and τ – time constant. For most tissues the parameters *k* and τ are usually unknown, but can be measured [2]. In addition to linear displacement, a shear wave is generated at the focal spot and propagates outwards with the speed depending on the tissue mechanical properties. MR elastography studies the propagation of these shear waves [7-9].

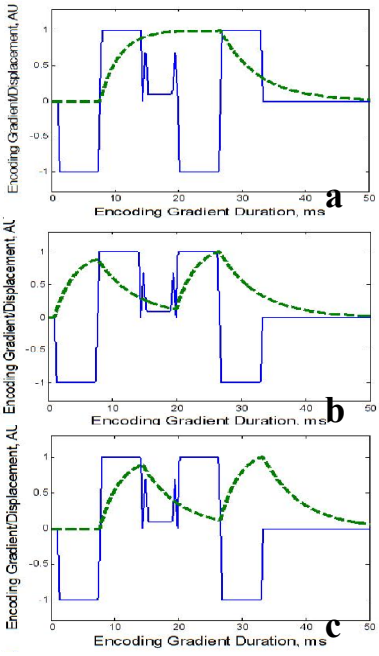


Figure 1. Encoding gradient configurations (solid line) and corresponding tissue displacement (dashed line) for **a)** repeated bipolar (RB), **b)** inverted bipolar 1 (IB 1) and c) inverted bipolar 2 (IB 2) cases.

Methods

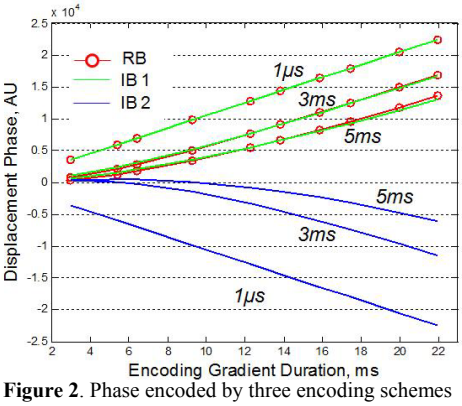
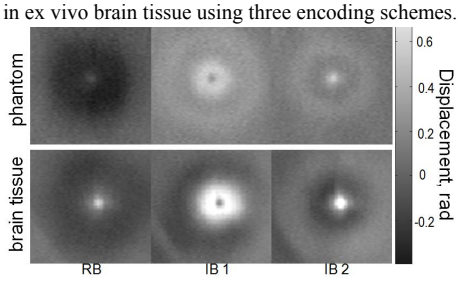


Figure 2. Phase encoded by three encoding schemes for nearly instantaneous tissue response and for tissue constants of 3ms and 5ms.

Figure 3. Displacement measured in a phantom and in ex vivo brain tissue using three encoding schemes.



Simulation: Three encoding configurations (Fig. 1) were modeled using encoding, slice select and crusher gradient parameters that were used on the MRI scanner to encode displacement in the plane perpendicular to the propagation of FUS beam. Tissue displacement due to FUS was modeled according to equation (2), using tissue time constants from 1 μs to 5 ms for the case of $\tau_{rise} = \tau_{decay}$,. Phase corresponding to each case was calculated according to Eq. 1 for encoding gradients' durations ranging from 3 ms to 22 ms. FUS duration for IB 1 and IB 2 schemes was equal to encoding gradient duration, and for RB case FUS pulse was longer by duration of refocusing pulse and the crushers.

Experiments: A 2D spin echo MR – ARFI pulse sequence was modified to allow for all three encoding schemes. A planar 1024-element transducer (f = 550 kHz, InSightec Ltd.) was used to create displacement in a phantom and in *ex vivo* porcine brain tissue using acoustic power of 56W. Displacement was imaged in coronal plane using a 3T GE MRI scanner and a solenoid breast RF coil (FOV = 16 x 12 mm^2, 256 x 96, BW = 15.63 kHz, TE/TR = 41/500ms). Duration of one encoding gradient lobe was 6.1ms, and time between the two bipolars was 5.6ms.

Results

The results of the phase simulation (Fig. 2a) show that for nearly instantaneous tissue response all three gradient configurations encode the same amount of phase (negative phase for IB1). For τ of several milliseconds, RB and IB2 show comparable phase with IB 2 outperforming RB for $\tau \le$ 3ms and RB approach taking over for higher tissue constants. Coronal images of displacement at the focal spot obtained with all three schemes in a phantom and brain tissue are shown in Figure 3.

Conclusion

In this work three displacement encoding techniques were evaluated experimentally and numerically. To select a particular encoding scheme based on phase simulation, the knowledge of the tissue response constants is required. Increase in τ leads to greater loss of phase due to undesirable encoding of displacement during relaxation. If the tissue constant is known, the timing of encoding could be manipulated in order to reduce these losses [4, 5]. In addition to the magnitude of encoded phase, the overall appearance of displacement is different for the three encoding techniques (Fig. 3). In both inverted bipolar schemes two FUS pulses are emitted per TR in comparison to one FUS pulse in repeated bipolar case, therefore generating a higher frequency shear wave apparent in Figure 3. The extent and the amplitude of the shear wave captured by the encoding gradient depend on tissue mechanical properties and attenuation coefficient as can be seen from the difference in displacement appearance between the phantom and the brain tissue.

References:

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