Towards Improving Predictive Capabilities of MR-ARFI for Transcranial Focused Ultrasound Therapy

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Introduction In recent MR-guided focused ultrasound (MRgFUS) treatments of neuropathic pain and essential tremor[1,2] , phase aberrations of the ultrasound caused by a patient's skull are estimated from the thickness and density of the skull derived from a computed tomography (CT) head image. Alternatively, MR-guided acoustic radiation force imaging (MR-ARFI) has been recently proposed to drive adaptive aberration correction algorithms[3-5]. MR-ARFI detects displacement of tissue caused by acoustic radiation force in the direction of encoding. In previous studies, the displacement was visualized along the axis normal to the face of transducer[3-5], further referred to as the main axis. In a hemispherical transducer (Fig 1A), however, the component of the acoustic force parallel to the main axis (Fig 1B) depends on the position of the element. Therefore, encoding only in this direction may be less sensitive to the ultrasound emission and phase aberrations of the peripheral elements of the hemisphere.

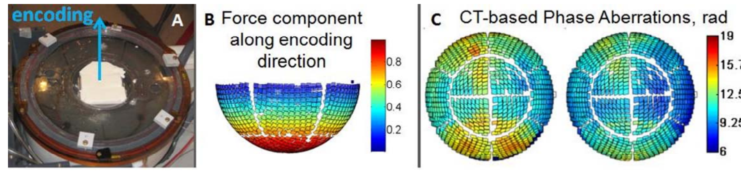


Figure 1. A. Ultrasound transducer with a tissue mimicking phantom placed at geometric focus. **B**. Components of radiation force from each element parallel to the encoding direction shown in A. **C**. Phase aberrations for transducer elements based on CT-derived estimation of two human skull acoustic parameters.

Purpose 1: to study experimentally the effect of electronically-applied human skull phase aberrations on the focal spot using MR thermometry that is independent of the direction of acoustic force, and MR-ARFI with encoding along the main axis of the hemisphere. **Purpose 2:** to measure the displacement phase from the individual groups of side elements, plates, using MR-ARFI with displacement encoding in oblique direction perpendicular to the face of the plate.

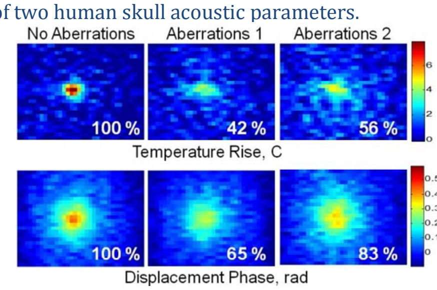
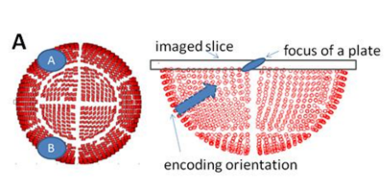


Figure 2. Cropped images of the focal spot without aberrations and with two sets of aberrations applied to the transducer.

Methods A tissue-mimicking phantom was placed into the geometric center of a hemispherical 720 kHz transducer (InSightec Ltd.) filled with degassed water and positioned inside a 1.5 Tesla MRI scanner (Fig 1A). Phase aberrations (Fig 1C), obtained previously during transcranial MRgFUS treatments in two patients[1] and estimated from skull CT images, were added electronically. Imaging was performed in the coronal plane where the geometric focus lies. The temperature rise was measured during a 10 sec sonication with 45W electric power (multi-phase 2D SPGR, TE/TR = 12 ms/15 ms). Displacement was measured using 2D spin-echo MR-ARFI with repeated bipolar encoding[6] (ultrasound pulse: 100W, 19 ms; TE/TR = 42 ms/500 ms). All images had a FOV of 30x24 cm^2, matrix size 256x102, slice thickness 6 mm. The average temperature and displacement were calculated for the cases of no aberrations and aberrations from subjects 1 and 2. To visualize the displacement from the individual plates, only the plates A or B were turned on (Fig 3A). Encoding gradient was applied along the direction normal to the face of a plate (Fig 3A). Displacement images were acquired with and without aberrations set 1.

Results Figure 2 shows the temperature rise and displacement phase images of the focal spot with and without aberrations applied. The temperature increase at the focal spot for the two aberration cases was 42% and 56% of the temperature rise without aberrations. However, on the displacement images the focal displacement higher fraction of the non-aberrated case: 65% and 83%. Figure 3 shows the displacement phase images of the ultrasound beam emitted from the individual plates A and B with displacement encoding in oblique direction normal to the face of each plate. Figure 3B shows that elements of plate B experienced stronger aberrations than elements of plate A confirming separately-simulated intensity predictions of the beams emitted from plates A and B.

Conclusion The results of this study experimentally demonstrated the difference in estimating the effect of the phase aberrations on the beam focus when using MR-ARFI and encoding displacement in direction of the main axis of the hemisphere versus using thermometry that is independent of individual elements acoustic force directions. In addition to the effect of encoding directionality, shear wave propagation also contributes to the difference between the temperature and displacement. In order to predict the effect of phase aberrations on all elements of the transducer more accurately, oblique displacement encoding was demonstrated in this work for two side plates of the hemispherical transducer. It revealed that one plate of the transducer was experiencing more aberrations than another plate. Performing MR-ARFI with oblique encoding for every plate of a hemispherical transducer is expected to provide more complete information about the phase aberrations due to patient's skull than encoding displacement only along the main axis of the hemisphere. In conclusion, it is important to account for the more complex geometry of the hemispherical transducer in order to use MR-ARFI effectively in phase aberration correction during transcranial MRgFUS treatments.



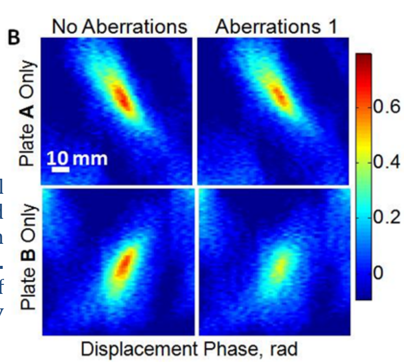


Figure 3. A. Schematic of hemispherical transducer showing the elements grouped into "plates". Slice position and direction of displacement encoding are shown. **B.** Obliquely encoded displacement images of the ultrasound beam focus emitted by plates A, B without/with aberrations.

References: 1. D. Jeanmonod et al., Neurosurg Focus, 2012. [2]] J.Elias Congres of neurological surgeons 2011 [3] B.Larrat et el. IEEE UFFC 2010 [4] Y.Hertzberg et al. Med.Physics 2010. [5]. L.Marsac et al. Med Phys 2012. [6] E. Kaye et al. MRM 2012.
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