Simultaneous Acoustic Radiation Force Imaging and PRFS Thermal Monitoring at 3T for MRgHIFU focusing

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Introduction

One challenge in MRgHIFU is to provide safe and thermally neutral focusing of HIFU beam pattern using acoustic radiation force imaging (ARFI). The radiation force is localized and highly directional (along the main propagation axis of the HIFU beam) while negligible outside the focal zone. This force initiates a tissue displacement correlated to the amplitude of the acoustic field and thus a phase shift that can be encoded in the MR signal using a motion encoding gradient (MEG) [1]. In addition, ARFI also provide 'stiffness weighted' images that may allow one to assess for pre- versus post- therapy changes in tissue. Since HIFU also causes tissue heating, temperature elevation and RFI effects are always associated, at various degree. We propose here to obtain a precise localization of the HIFU focal point by subtracting GRE phase images from two independent acquisitions, where ARF-induced phase shift is sequentially encoded with positive and, respectively, negative monopolar MEG pulse. For illustration, the MEG was implemented here along the slice-select direction.

Materials and Methods

A FLASH sequence was modified to integrate positive or negative MEG in the slice- select direction (max amplitude=25mT/m, slew rate (200T/m/s), duration= 6ms). The total zero-momentum of gradient along the slice-select direction was compensated to null value. The HIFU burst was produced by a 256 multi-element transducer (Imasonic, Besançon, France). Natural focal lenght and aperture of the transducer are R=130mm and respectively d=140mm (f = 974 kHz). The MR sequence generated an optical trigger at the end of the slice refocusing gradient. A home made optic-to-TTL conversion & timer board provided the sonication window for the HIFU generator (IGT, Pessac, France). The HIFU burst was required to the generator δ=1ms before the MEG and its duration was set to $\Delta t = \delta+\tau$ = 7ms. The acoustic power was set each time to P_a=196W (pre-calibrated by balance measurements). Coronal images perpendicular to the HIFU beam were acquired on a 3T MR system (TIM Trio, Siemens AG, Germany). Main imaging parameters were: voxel = 1x1x5.0 mm, *TR/TE/FA* = 100ms/18/35, loop coil (11cm). Two experiments were performed ex vivo (degassed Turkey muscle): 1) full FOV and full k-space (FOVz=FOVx=128mm, 128 acquired lines in total, 12.8s/image) and 2). reduced phase FOV (FOVz=65%) and 75% partial Fourier (PF) (63 acquired lines in total, 6.3 s/image).

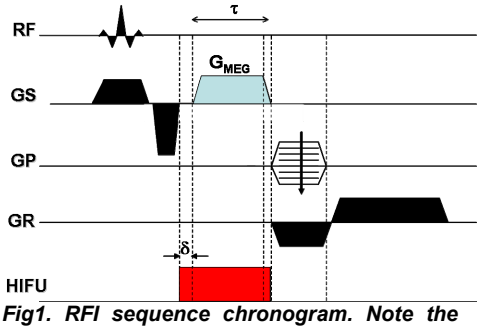
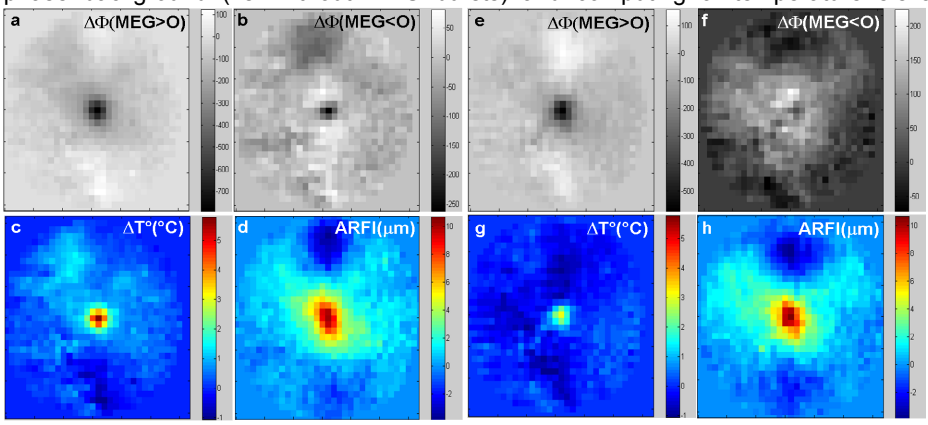


Fig1. RFI sequence chronogram. Note the 6ms MEG along the slice-select direction. The (+) polarity is shown. Spoiling gradients are not shown. Some time overlapping gradients were split for illustration purpose.

Note the physical spatial resolution was the same in both experiments and the reduced FOV still largely covered the ARFI contrast region around the focus. Temperature (MRT) and ARFI maps were obtained by performing the half-sum and half-difference respectively of the phase-shift images generated with (+) and (-) MEG-polarity. For a monopolar rectangular gradient, the maximum tissue displacement by radiation force can be calculated [2,3] from $\Delta y=\Delta\Phi/(\gamma \cdot G_{MEG} \cdot \tau)$ where $\gamma/2\pi$=42.58MHz/T. The trapezoidal MEG (slew rate = 200T/m/s and rising time=150µs) was approximated as a rectangular one. Phase unwrapping, temporal subtraction of reference phase background (i.e. without HIFU bursts) and computing of temperature elevation at each pixel were performed in Matlab.



FOVz=FOVx, matrix 128x128 **FOVz=65%FOVx, matrix 84x128, Partial Fourier=6/8,**

Results

In the first experiment (Fig.2, left), the measured displacement at focus was Δy=10.23µm whereas the measured T° elevation at focus was ΔT_{max}=5.78°C. In the second experiment (Fig.2, right), Δy=10.71µm and ΔT_{max}=3.07°C.

Fig2. Phase shift as obtained after subtraction of reference phase background (i.e. no HIFU bursts) for positive (a) and negative (b) MEG polarity. ARFI (c) and T°(d) maps. Displayed FOV = 32x32mm.

Discussion and Conclusion

Our displacement values in tissue are in agreement with previously published data [1-3]. By reducing the number of acquired k-space lines from 128 (FOV_z=FOV_x, full k-space) to 63 (FOV_z=65% FOV_x and PF=75%) we reduced the number of synchronized HIFU shots, i.e. the localized energy deposition, hence ΔT_{max} was reduced by a factor of 1.88 that is equal to the predicted factor (128/63=2.03) within the noise SD. This also improved by a factor of 2.03 the temporal resolution, without penalty on spatial resolution nor on ARFI CNR. The balance between ARFI contrast and energy deposition depends on the number of MR excitation RF pulses (equal to the number of HIFU bursts per image since HIFU burst is triggered by the MR sequence). Thermal monitoring simultaneous to ARFI is a valuable safely tool. To date, using a simple PF-GRE-ARFI sequence, ARFI can localize in seconds HIFU focal spots with sufficient CNR (≈11 at focus) and low T° elevation (+3°C). However, a CNR similar to ARFI was obtained by PRFS MRT alone, considering for instance in situ optimization of focusing. Nevertheless, the current sequence measures tissue displacement hence tissue stiffness, allowing one to investigate pre- and post- therapeutic tissues changes. Energy deposition can be further reduced (while maintaining identical CNR in ARFI), using even more efficient k-space filling strategies such as GRE-EPI kernel and/or parallel imaging. Further work should address such improvements.

References: (1) Souchon et al. MRM 2008. (2) Plewes D.B. at al. JMRI 1995. (3) McDannold et al. MedPhys 2008.