ABSOLUTE MR THERMOMETRY FOR NEAR-FIELD MONITORING DURING MR-HIFU HEATING

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Introduction

In MR-guided high intensity focused ultrasound (MR-HIFU), MR temperature mapping allows for precise control of the thermal dose in the targeted region (1). However, when applied to abdominal organs, tissues in the near-field region of the HIFU beam are at risk of undesired heat accumulation (2). Therefore, there is a need for temperature mapping in the near-field to visualize the cumulative effects of subsequent sonications. In this work we demonstrate the use of multi-gradient echo (mGE)-based absolute MR thermometry (3) at the fat-muscle interface for the assessment of the thermal build-up in the near-field area.

Methods

Theory: the proton resonance frequency of water is temperature dependent, whilst it is temperature independent for fat (4). In the presence of both water and fat their difference resonance frequency Δf_{wf} can be used to calculate the absolute temperature, using: $T = \frac{1}{\alpha}\left(\Delta\delta_{wf}(T_{ref}) - \frac{\Delta f_{wf}}{\gamma B_0}\right) + T_{ref}$ [1], where α is the electron screening thermal coefficient of water and $\Delta\delta_{wf}$ is the chemical shift between water and fat at a certain reference temperature T_{ref}. Values used in this study: $\Delta\delta_{wf}$=3.2ppm at T_{ref}=37˚C and α=-0.01 ppm/˚C (4,5). As the mGE modulus signal in voxels containing both water and fat shows an oscillation frequency equal to Δf_{wf}, the value of this parameter can be retrieved by voxel-wise fitting of the mGE signal equation to the acquired data points. Subsequently, the absolute temperature can be calculated (3). The following measures were taken to make the fit procedure more robust. First, an estimate for the signal amplitude directly after excitation (at t = 0), A_0, was obtained by extrapolating the highest amplitude value with a single exponential. Second, as unpublished simulations showed that using slightly incorrect R_2^* values has little effect on the accuracy, the simplification $R_{2,w}^* = R_{2,f}^* = R_{2,mean}^*$ was made, where $R_{2,mean}^*$ is the mean of measured $R_{2,w}^*$ and $R_{2,f}^*$ values. Now, the mGE signal can be modeled as: $S(t) = A_0 e^{-2R_{2,mean}^* t}\sqrt{\lambda^2 + (1-\lambda)^2 + 2\lambda(1-\lambda)\cdot\cos(2\pi\Delta f_{wf} t + \Delta\varphi)}$ [2], where 0≤λ≤1, representing the relative signal amplitude for one component and $\Delta\varphi$ is the initial phase difference. To extract Δf_{wf} values, a nonlinear least-squares trust-region fit procedure (Mathworks,Natick,MA) was used. Voxels were excluded when R^2<0.75 and/or when the fit process ended at a boundary value for λ of 0.25 or 0.75. Absolute temperatures were calculated using Eq.1.

Experimental set-up

Validation: A fresh ex vivo porcine abdominal wall sample was heated in a waterbath and scanned during cooldown. A fiber-optic temperature probe was inserted into the center of the sample. To include both water and fat, the imaging slice was positioned at the muscle-fat interface (Fig.1). mGE images were acquired with a head RF receive coil with the following scan parameters: ETL=32, TE_0=1.4ms, ΔTE=1.3ms, TR=52.5ms, voxel size 2x2x8mm^3, FOV 130x130mm^2. The scan duration was 4s per dynamic. Additional scans with ProSet fat and water suppression were used for $R_{2,w}^*$ and $R_{2,f}^*$ measurements in the muscle and fat, respectively (Fig 1).

Monitoring of HIFU heating: To demonstrate the ability to detect local HIFU heating, an experiment was performed using a clinical MR-HIFU system (Philips Sonalleve,Vantaa,Finland) on a fresh ex vivo porcine abdominal wall sample. A focal spot of 2 mm was placed in the imaging slice and we sonicated with 50W for 30s. A standard clinical MR-HIFU pelvis RF receive coil was used and scan parameters were as described above, except for: voxel size 2.8x2.8x8mm^3, FOV 180x180mm^2.

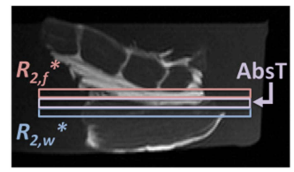

Figure 1 Slice positioning for absolute thermometry (AbsT) and $R_{2,w}^*$ and $R_{2,f}^*$ measurements.

Results

The mean measured R_2^* values over a temperature range of 25°C to 55°C were $R_{2,w}^*$=28s^{-1} and $R_{2,f}^*$=31s^{-1}, thus $R_{2,mean}^*$=29.5s^{-1} was used in the fitting procedure. Absolute temperature maps of the cooldown experiment are shown in Fig.2a. Probe temperatures are given in the images and the probe location is indicated by a square. Probe temperatures were quantitatively compared with the mean calculated temperatures of an ROI of 5x5 voxels near the probe (Fig.2b). Results from the HIFU heating experiment are shown in Fig.3.

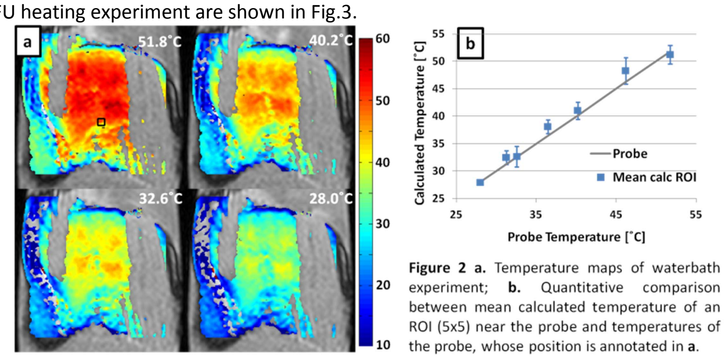

Figure 2 a. Temperature maps of waterbath experiment; **b.** Quantitative comparison between mean calculated temperature of an ROI (5x5) near the probe and temperatures of the probe, whose position is annotated in **a**.

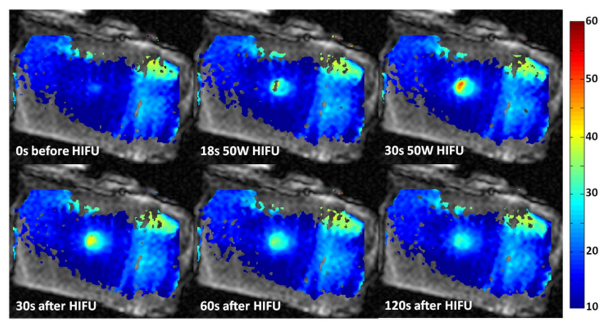

Figure 3 Temperature maps from the local heating experiment with HIFU.

Discussion

We have shown that mGE-based absolute thermometry can be used for measuring the temperature at the subcutaneous fat-muscle interface in the abdomen. The temperature maps show the expected spatiotemporal cooldown patterns, starting at the outer rim (Fig.2a). In the quantitative comparison (Fig.2b), we observed a close correspondence between the probe temperatures and the calculated temperatures. The results of the local heating experiment (Fig.3) clearly show local heating and cooldown, although the temperatures on the outer part of the tissue sample seem to have been consistently overestimated. This artifact is subject to our current investigations. Note that the top-left image shows a slight temperature elevation at the focal point before HIFU. This is most probably a result of the test sonication performed prior to the actual sonication to verify the spatial accuracy of the HIFU beam. Note that conventional PRFS-based MR thermometry, which calculates temperature changes between subsequent images, cannot visualize such effects. This observation clearly illustrates the potential of this method to detect cumulative heating in the near field region due to subsequent sonications.

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References: (1) De Poorter et al., Magn Reson Med, 1995, 33: 74-81. (2) Payne et al., Med Phys, 2011, 38(9): 4971-4981. (3) Sprinkhuizen et al., Magn Reson Med, 2010, 64: 239-248. (4) Kuroda et al., Int J Hyperther, 2005, 21(6): 547-560. (5) Ren et al, J Lipid Res, 2008, 49: 2055-2062.