Using a simple thermal model to correct errors in PRFS MR Thermometry due to heat-induced tissue susceptibility changes

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Introduction: During MR-guided high intensity focused ultrasound (MR-HIFU) ablation of fat-containing tissues like the human breast or liver, heat induced susceptibility changes may give rise to local field disturbances, leading to errors in PRFS thermometry of the water-based tissues [1]. Because the error is small compared to the actual temperature change [1], the initial uncorrected measured temperature change combined with a thermal model may be sufficient to estimate the field disturbance and thus correct for these errors. In this work, we demonstrate the feasibility of a correction method based upon this principle in a HIFU experiment.

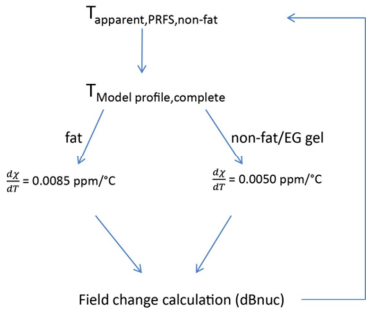

Fig. 1) Flow diagram of the correction method

Methods: Correction method: Fig.1 shows an overview of the correction method. The initial uncorrected PRFS thermometry is used as an input to a thermal model to estimate the temperature change in the adipose and water-based tissues. The susceptibility versus temperature coefficients (dχ/dT) of both tissues are then used to determine the susceptibility change. The change in susceptibility is then used to calculate the change in nuclear magnetic field using a memory efficient and fast Fourier-based algorithm [2]. The initial PRFS temperature is then corrected based on the known field change and the steps are repeated until the corrected temperature change remains stable (defined here as temperature changes under 0.1°C). Experiment: The correction method was demonstrated in a phantom containing porcine adipose tissue and an ethylene glycol (EG) gel to represent the water-based tissue. Ethylene glycol was used since it provides two resonances, one of which has a temperature dependent electron screening constant. This allowed us to separate field and temperature effects, as will be explained below. The EG gel, with a dχ/dT of 0.0050 ppm/°C, consisted of 2/3 ethylene glycol, 1/3 deuterium oxide, 3% agar, and 3% silica. The experiments were performed on a clinical MR-HIFU system (Sonalleve, Philips Healthcare) integrated into a 1.5-T MR Scanner. A cylinder was half filled with porcine adipose tissue and half with EG gel (Fig. 2a). As the MR receive coil a 4.7 cm diameter microcoil was used which was positioned close to the sonication focus and imaging slice. The EG gel/fat interface was sonicated with a 4-mm diameter volumetric cell using 20W of acoustic power during 60s. During sonication, 2D multi echo spoiled gradient recalled (SPGR) scans were acquired dynamically. The imaging slice was positioned perpendicular to the HIFU beam and intersected the HIFU focus. The imaging parameters used were: FOV=128 x 128 mm, matrix=128x128, slice thickness = 5 mm, NSA=1, 8 echoes, TR=45 ms, TE0=1.65 ms (first echo time), ΔTE = 3.09 ms (echo spacing), 40 dynamics.

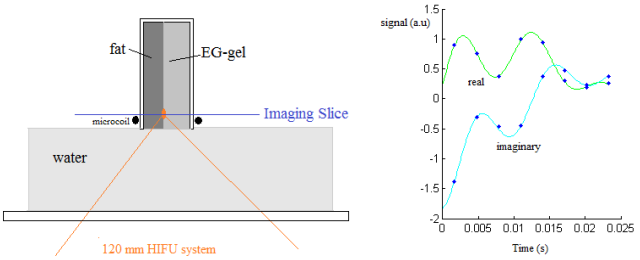

Fig. 2 a) Experimental setup of the HIFU sonication of the fat/EG-gel interface and b) example fits of the real and imaginary signal.

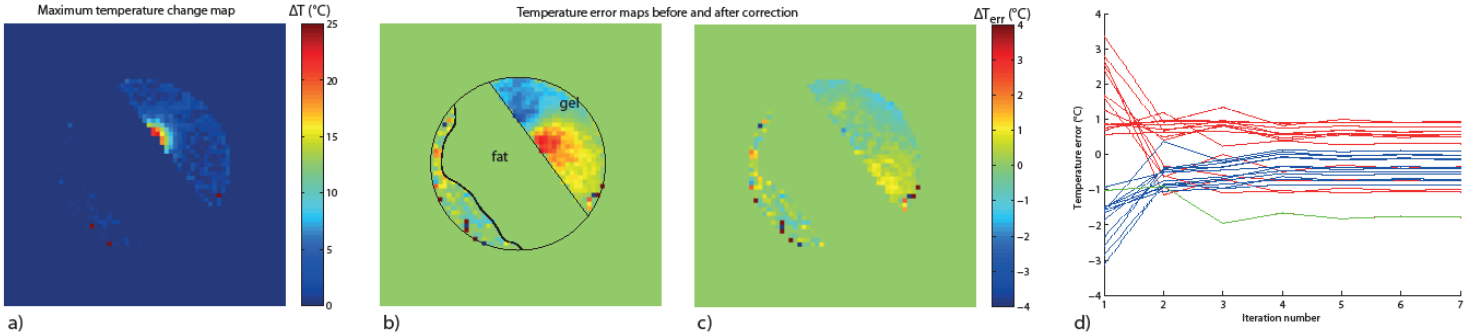

Fig. 3) a) Maximum absolute temperature change map. Temperature error b) before correction, c) after correction and d) as function of iteration step (1=before correction) for the EG voxels at the fat-EG gel interface.

Post-processing: For the EG-gel region, the signal was fit to a two peak model in the time domain [3] using code written in MATLAB (fig. 2b). The signal was first normalized and the least square difference between the model signal and data was minimized using the Levenberg-Marquardt algorithm. In total nine variables were used: the amplitude (A), R2* relaxation time, frequency (f), and initial phase (φ_0) for both the CH_2 and OH peaks. Additionally, the variable g represented the degree of Lorentzian or Gaussian line form. The starting values used for the fitting algorithm were [A_{CH2} $R2^*_{CH2}$ f_{CH2} $\varphi_{0\,CH2}$ A_{OH} $R2^*_{OH}$ f_{OH} $\varphi_{0\,OH}$ g] = [1, 40 1/s, 15 Hz, 0, 1, 40 1/s, 100 Hz, 0, 1]. The shift in CH_2 position represented the field disturbance. The shift in OH peak position represented the uncorrected PRFS thermometry. The absolute temperature change was calculated using the change in frequency separation between the OH and CH_2 peaks. The temperature coefficient of -0.01 ppm/°C was used to convert peak shift to temperature change. The correction method (Fig. 1) was then applied by fitting the uncorrected PRFS temperature change with a 2D Gaussian function.

Results: The maximum absolute temperature change in the EG-gel was 22.4°C (Fig. 3a). Fig. 3b and c show temperature error maps before and after correction. The maximum errors of -3.1°C and 3.4 °C (found at the gel/fat interface) before correction were reduced to -0.7 and 0.9 °C respectively. Figure 3d shows the temperature error for the EG-gel voxels located along the edge of the gel/fat boundary as function of iteration step. Six iterations were found sufficient for the correction procedure to stabilize. With the exception of one voxel the temperature error decreased to less than 1°C after the correction.

Discussion & Conclusion: The correction method was shown to be feasible in an adipose/EG-gel phantom. The method may further be improved by using Wiedemann's additive law of susceptibility [4] for the voxels with mixed tissue types and by applying a more realistic thermal model. For the application in fat-containing tissues, water/fat separated images may be used as input to the correction method. Although the method reduced the error and converged for the current experiment, further research will investigate the application in tissues with more complex water/fat distributions and heating profiles.

References: [1] Sprinkhuizen et al., Magn.Reson.Med.;64(5):1360-72, 2010, [2] Bouwman et al., Magn.Reson.Med.;68(2):621-630, 2012, [3] Van Hamme et al., J. Magn Reson; 129(1):35-43, 1997, [4] Mulay LN. Magnetic susceptibility. New York: John Wiley & Sons; 1963