Correcting for B0 Field Drift in MR Temperature Mapping with Oil References

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Introduction Hyperthermia has been shown to be highly valuable as an adjunct to radiation therapy in such cases as recurrent cancer in the chest wall [1]. Accurate tumor and normal tissue temperature measurement is a key factor for successful treatment. Invasive thermometry provides accurate but spatially limited measurements. Regional temperature mapping via MR methods potentially can increase accuracy of control of the heated region. Previous work has shown the value of using the temperature sensitivity of the tissue water proton resonant frequency shift (PRFS) [2]. However, main magnetic field frequency drift over the treatment duration can cause errors in frequency shift approaches if not corrected.

Theory The proton chemical shift of water is temperature sensitive. For a gradient echo sequence in a stable scanner voxel phase will change over time as $\Delta\phi = \phi(T_1) - \phi(T_0) = -\gamma T_E \alpha \Delta T B_0$ where γ is the gyromagnetic ratio, T_E is echo time, and *α* is the temperature coefficient constant (0.01 ppm/°C) of water. As shown, this method is confounded by changes in the main magnetic field (δB_0) due to other factors than temperature. By comparison, oil does not have temperature dependent chemical shifts and can be used to measure δB_0 changes due to non-temperature effects.

Methods The PRFS protocol acquired gradient echo images, TE=20ms, TR=34ms, BW=32kHz, 128x128 pts, FOV 30cm, 4 slices, 7mm thick, approximately every 60 seconds for the duration of the experiment on a 1.5T GE system (GE Healthcare, Waukesha, WI). MR temperature measurements were made in a cylindrical acrylimide phantom (5x14 in) and, after IRB approval, a patient with a leg sarcoma during treatment. Luxtron fluorescent probes in catheters inserted into the phantom/tissue measured temperatures during image acquisition. Phantom/tissue was heated in a mini annular phased array [3] with 4 RF antennas coupled through a water bolus sleeve (see Fig. 1A,D) at 140 MHz. Complex data were stored and transferred offline for processing using in-house software developed in IDL (ITT-VIS, Boulder, CO). Four silicon oil reference markers bracketed objects of interest. Oil phase change with time was measured within these locations. Global δB_0 was estimated using the IDL MIN_CURVE_SURF routine to fit a minimum curvature spline surface [4] to the oil reference locations. Non-temperature related phase shifts were subtracted from the original data.

Results Fig. 1 (A,D) Magnitude images of the first time point for the phantom and leg sarcoma. Temperature change maps (B,E) contain 3 and 2 voxels, respectively, with temperature plots in (C,F). Both uncorrected and δB_0 corrected temperature estimates are plotted for all color-coded voxels. Luxtron values are plotted in black. Uncorrected temperatures in the phantom varied up to 17 °C and up to 8 °C in the leg. Temperatures in ROIs aligned with Luxtron probes showed excellent agreement after δB_0 correction.

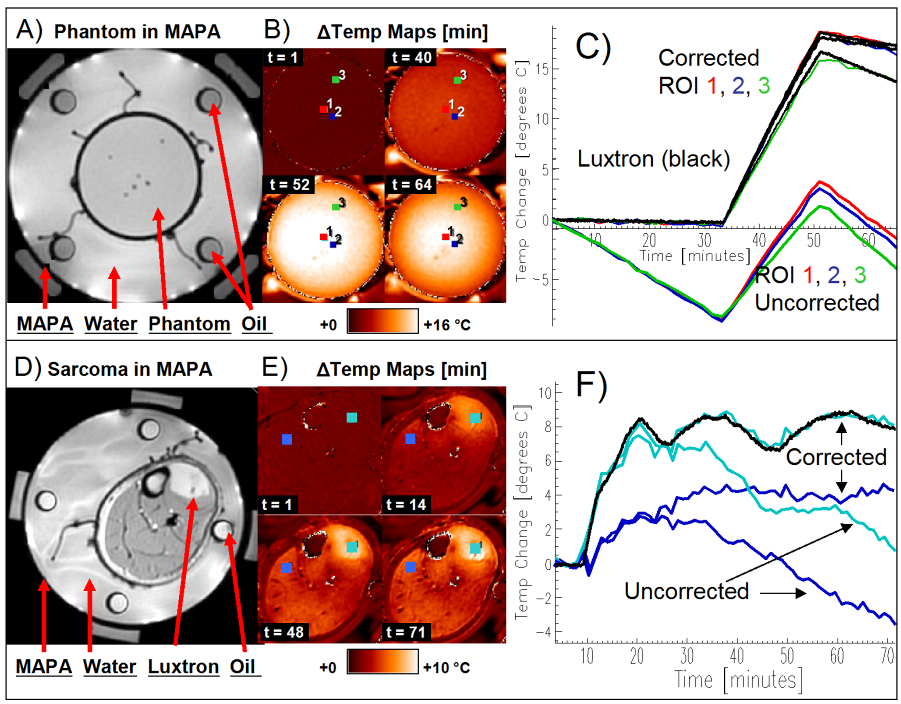

Figure 1. Images, temp maps and ROI plots for phantom and leg.

Discussion High signal to noise and image stability were achieved in the phantom study. SNR and temperature calculation in the sarcoma patient suffered from biological noise and patient movement. The use of four markers gave reasonable results for the δB_0 changes seen in this study (max ~70 Hz/hr). Additional and more closely spaced markers, relative to the object being imaged, might be necessary to account for more complicated δB_0 changes. However, reference numbers and placement must not interfere with RF heating elements. One possible additional source of oil signals would be to make use of lipids in subcutaneous fat, via a water-fat separated imaging method.

References and Acknowledgements 1. Jones EL, etal, Clin Cancer Res 10, 4287-93, 2004. 2. Carter DL etal, Int J Rad Onc Biol Phys 40, 815-22, 1998. 3. Zhang Y, etal, IEEE Trans Biomed Eng, 40, 780–7, 1993. 4. Franke R, Comp Math Appl, 8, 273-281, 1982.

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