MR Temperature Imaging using an Aliased Fat-Water Fast Chemical Shift Acquisition

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Introduction

Noninvasive measurement of the spatiotemporal temperature distribution using magnetic resonance temperature imaging (MRTI) based on the temperature sensitivity of the water proton resonance frequency (PRF) shift [1] is well established and has been shown to provide useful temperature feedback for control and dosimetry of thermal therapies in a variety of anatomical sites [2,3]. A fast phase-difference technique [4] has been the preferred method for estimating temperature. Limitations of this approach include sensitivity to intravoxel lipid contamination, inter- and intra-scan motion, susceptibility and, during extended monitoring times, field drift. In order to expand the utility of this technique to anatomical regions where these limitations limit the ability to obtain robust, accurate temperature estimates during therapy, some investigators have turned to frequency mapping using multi-echo sequences or fast chemical shift imaging (CSI) techniques [5,6] as a solution that simultaneously minimizes the errors associated phase-difference imaging and provides the possibility of using lipids as an internal reference.

While the fast CSI techniques address many of the limitations of the phase-difference technique, CSI acquisitions are still limited by spatiotemporal resolution versus SNR and spectral resolution tradeoffs [6]. However, for PRF based temperature imaging, it's important to note that accurate measurement of peak heights and widths are not really a requirement of the sequence, only the relative placement of the peaks. Therefore, the echo-spacing constraints imposed on an echo-train based acquisition can actually be relaxed in favor of more SNR or increased temporal efficiency as long as the aliasing of fat and water isn't made so severe as to overlap the peaks and inhibit the ability to resolve them (Figure 1). This is particularly important at higher fields where much stronger gradients are required to adequately reduce echo-spacing and therefore temporal resolution is often sacrificed for a multi-shot approach to overcome the gradient limitations. In the current investigation, we investigate the feasibility, advantages and disadvantages of using a fast CSI technique based on a gradient-echo train acquisition with intentional aliasing of the lipid peak as a novel method for rapid, self-referenced MR temperature imaging.

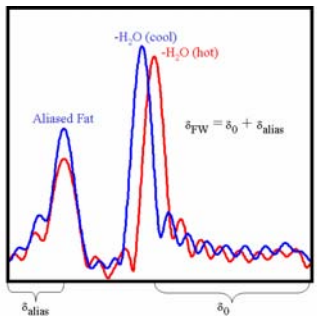

Fig. 1: Aliased fat-water spectrum from fast CSI.

Materials

All imaging was performed on a 3.0T whole body MR scanner (TwinSpeed EXCITE HD, GE Healthcare, Waukesha, WI). A phantom composed of heavy whipping cream was heated to 60ºC and was imaged using an 8-channel receive only phased-array head coil (MRI Devices Corp, Gainesville, FL) during cooling. Temperature in the phantom was monitored using a fluoroptic temperature monitoring system (Luxtron, Santa Clara, CA). A multi-echo 2D fast gradient-echo sequence was used to collect 16 echoes with an echo spacing of 1.8 ms, TR of 75-ms, flip angle of 30º, readout bandwidth of 651 Hz/pixel, acquisition matrix of 128 x128, field of view of 30-cm x 30-cm. Further, a parallel imaging acquisition technique (ASSET) was used in the phase-encoding direction to reduce acquisition time (acceleration factor of 2). With this particular choice of sequence parameters, the sequence gave a spectral bandwidth of 556 Hz and spectral resolution of 35 Hz/point.

Results

The complex phase-difference was calculated from the longest echo-time (29.7-ms) to approximate a "lipid suppressed" PRF shift from water as the majority of lipid signal should have decayed away at 3T owing to the shortened T2*. As shown in Figure 2, the phase-difference technique in phantom tracks extremely well with the measured frequency shifts from the fast CSI acquisition. However, both deviate from the fluoroptic probe measurements over time, presumably from magnetic field drift, thus both techniques return similar temperature sensitivity coefficients (-0.0085 ppm/ºC for the phase-difference and -0.0089 ppm/ºC for fast CSI). However, by using the fat peak in the phantom as an internal reference, the estimated temperature from the fast CSI technique appears to compensate for the field drift extremely well (Figure 2) resulting in a much better correlation to the fluoroptic probe measurements and, as seen in Figure 3, results in a temperature coefficient (-0.0101 ppm/ºC) in line with previous measurements [1].

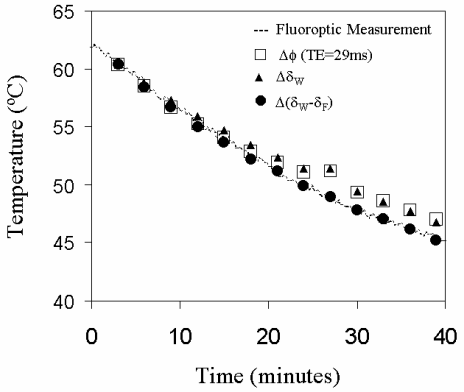

Fig. 2: Measured temperature during cooling in fat-water phantom versus time with various MRTI estimates (note an assumed temperature sensitivity coefficient of -0.01 ppm/ºC.)

Discussion

As noted in previous work [5], there were problems tracking the fat peak above 50ºC. Unless corrected for in the acquisition as demonstrated in Ref [6], this would lead to problems with using intravoxel fat as internal reference in the region of heating for ablative purposes. Therefore, in these experiments, and in current in vivo experiments, we are developing an algorithm to find nearby lipid tissue to use as an internal reference. Preliminary data, in an in vivo prostate model without heating, show that this method may be useful for correcting small deviations. However, further in vivo tests are warranted to fully explore this. One of the most attractive features of the acquisition is that, while maintaining many of the benefits of a chemical shift MRTI acquisition, the spatiotemporal resolution is still acceptable for most thermal therapy uses. With the parameters used here, two interleaved planes can be acquired every 4.8 seconds. Using higher acceleration factors and incorporating a fractional Fourier acquisition in the phase-encode direction with a phase-preserving reconstruction, such as POCS, the TR could be extended to improve slice efficiency at 3T while simultaneously reclaiming some SNR lost from the acceleration schemes. At 1.5T, the bandwidth can be reduced with the relaxation of the sampling constraint for fat-water. Similarly, this SNR can be traded in for faster acquisition times using the aforementioned acceleration techniques. Investigations into how to optimize parameter selection are ongoing as is further extension of this work to in vivo models.

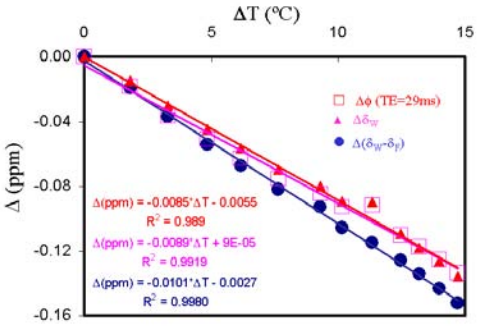

Fig. 3: Associated calibration curves for each measurement technique

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