Fast MR thermometry using phase referenced asymmetric spin-echo EPI for high field

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Introduction: The proton resonance frequency (PRF) shift MR thermometry method is inherently very sensitive to magnetic field perturbations in time [1]. Chemical shift imaging, RF saturation and inversion methods can correct for magnetic field changes using a temperature-independent reference substance (e.g. fat) [2,3]. The frequency selective spin-echo (SE) technique of Ivanov et al has been shown to provide high accuracy phase referenced temperature data [4,5], but this method has one drawback: It is not possible to acquire water and reference images at the same position closely in time. This can be achieved by the SE gradient reversal technique, which uses slice-select gradients of opposite polarity for frequency-selectivity [6,7]. To minimize the time between the acquisition of the water and reference images while retaining the maximum signal amplitude for both acquisitions a more conservative condition on the gradient reversal technique was used. Thus, the slices of the wanted (e.g. water) and unwanted (reference, e.g. fat) chemically shifted species do not overlap. Immediately after the acquisition of the water image, the reference image can be acquired at the exact same position as the water image with the full fat magnetization available for image formation (see Fig. 1). The interval between the water and reference image can therefore be reduced to about 50ms making it possible to correct for many causes of field drift like breathing and at least partially for cardiac pulsation. We tested this novel approach by temperature stability measurements in-vivo.

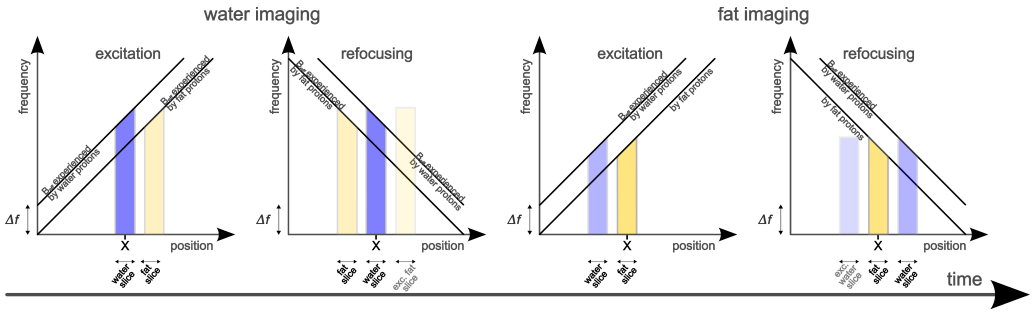

Fig. 1 During slice-selective excitation and refocusing, chemical species are displaced in slice direction depending on gradient amplitude and chemical shift of the species (for fat-water at 7T the frequency difference is Δf≈1000Hz). For the presented method, the excitation and refocussing slice position of the unwanted chemical species is displaced from the imaging position (position X) so far, that this species can be immediately imaged afterwards without loss in signal due to saturation.

Methods: All experiments were performed on a 7T whole-body MR scanner (MAGNETOM 7T, Siemens, Germany) using an 8-channel phased-array coil (RAPID Biomedical, Germany). The study was approved by the local ethics committee and all subjects gave informed consent. One to six coronal water slices and the same number of fat slices were acquired across the occipital lobe in 4 healthy subjects. To achieve good separation of fat and water, using the same slice thickness, the excitation slice-select gradients were -6.4mT/m and 6.4mT/m for the refocusing gradient [7]. The separate fat and water images at the same position were acquired by alternately switching the RF-pulses to the water and fat frequencies in immediate succession, within a repetition time of either 250ms or 2000ms, depending on the number of slices. The other imaging parameters were: TE=21ms, k-space centred at 33.6ms → effective echo time TE_{eff}=12.6ms, bw=1502Hz/Px, resolution 128×128, voxel size: 1.5×1.5×1.5mm^3, GRAPPA factor 4. The subcutaneous fat phase around the skull was fitted by a 2D linear function for each time-step and slice and then subtracted from the water phase of the brain to correct for field perturbations affecting the water phase. The corrected water phase images were then converted to temperature change maps according to the PRF phase shift method [1]. The modified actual flip angle technique was used to ensure that the flip angle achieved for the excitation pulse was not above 135° in the slices imaged and B_0 was homogeneous enough to ensure chemical selectivity of the SE sequence (below ±200Hz in each slice), determined by the Bloch simulation [8,9].

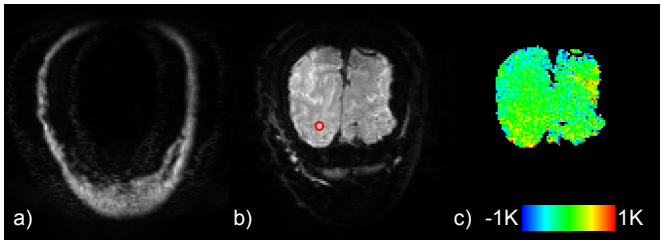

Fig. 2 (a) Fat and (b) water images of a representative slice with a TR=2s and the (c) unsmoothed referenced temperature change map for the last repetition after 10 minutes. The red circle in (b) marks the 9 voxel ROI used for Fig. 3.

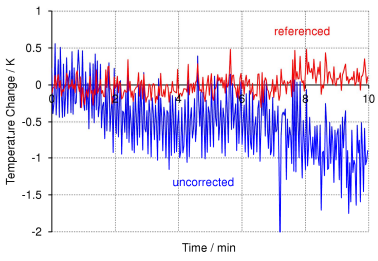

Fig. 3 Observed temperature time courses of the ROI depicted in Fig. 2(b) with a TR=2000ms.

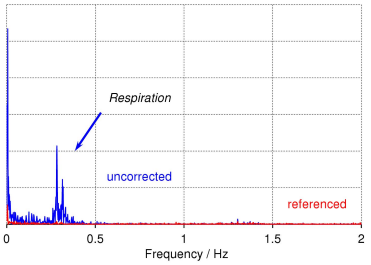

Fig. 4 Power spectrum of the uncorrected and referenced temperature time courses for the ROI of Fig. 2(b) with TR=250ms.

Results and Discussion: Fig. 2 shows the fat (a) and water (b) image of a representative coronal slice across the occipital lobe. The water and fat are fully suppressed, despite some minor ghosting in the fat image (a). Fig. 2(c) is the referenced temperature change map after 10min. Compared to the unreferenced average temperature change of ΔT=-1.1±0.5K, the corrected temperature map referenced by the fat images shows practically no temperature change (average ΔT=-0.1±0.3K)[10]. The time course in Fig. 3 shows that not only the drift over time is corrected, but the contributions from higher frequency fluctuations are also reduced. Fig. 4 displays the power spectrum for the temperature curves with the TR of 250ms. Several frequency components in the range from 0 to 0.5Hz are substantially reduced, such as the large respiratory peaks around 0.3Hz, and peaks between 1 and 1.5Hz probably due to cardiac pulsation [11]. In general, the fat correction improved temperature measurements for all time-series. For the inferior part of the brain the correction worked better, probably due to the larger amount of fat in close proximity to the brain. In the superior part of the most anterior slices the fat correction did not always improve the temperature measurement, probably due to the low fat signal within the slice. This affected 5 out of 64 slices. With more elaborate methods similar to Salomir et al the fitting of the fat phase might be further improved [12].

Using referenced MR thermometry with a high temporal resolution of 50ms between water and fat acquisition, and high temperature precision, may enable in-vivo monitoring of temperature changes due to RF power deposition, for individual real-time SAR monitoring.

References: [1] Rieke V et al, JMRI 27(2):376, 2008 **[2]** Kuroda K et al, MRM 38(5):845, 1997 **[3]** Taylor BA et al, MedPhys 35(2):793, 2008 **[4]** Ivanov D et al, MRM 64(2):319, 2010 **[5]** Streicher MN et al, #529 ISMRM 2011 **[6]** Park HW et al, MRM 4(6):526, 1987 **[7]** Volk A et al, JMR 71(1):168, 1987 **[8]** Allard P et al, JMR 129(1):19, 1997 **[9]** Amadon A et al, #1248 ISMRM 2008 **[10]** Collins CM et al, J Appl Physiol 97(6):2051, 2004 **[11]** Petridou N et al, MRM 27(8):1046, 2009 **[12]** Salomir R et al, IEEE Trans Med Imag 2011.