Non invasive monitoring of the brain temperature during mild hypothermia

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Introduction: Monitoring the brain temperature changes during mild hypothermia with high temporal and high spatial resolution is difficult task for magnetic resonance thermometry. Conventional single voxel spectroscopy [1] or spectroscopic imaging (SI) methods are unsuitable because of poor spatial resolution and long acquisition time to obtain acceptable signal-to-noise ratio of water-suppressed spectra. The weakness of phase-mapping techniques [2] is high sensitivity to inter-scan motion since the temperature changes are computed using the phase differences between successive MR images. In this work, SI with high spatial and reduced spectral resolution was used for the brain temperature mapping. The main purpose of this study was to verify the feasibility of the method. A secondary goal was determining the temperature coefficient of water chemical shift in the brain without internal spectral reference.

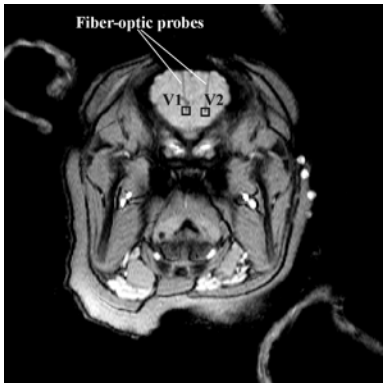

Fig. 1a

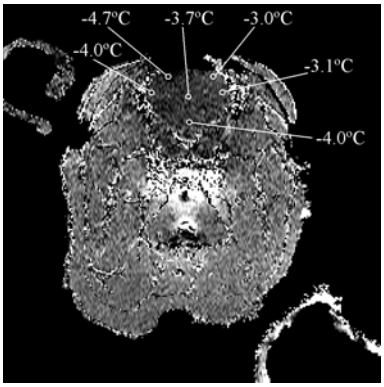

Fig. 1b

Materials and Methods: Six anesthetized pigs were used in this study. Selective brain cooling was performed through both the nasal cavities using water cooled balloon catheters [3]. Cooling medium at a flow rate of 100 ml/min was intended to produce brain hypothermia (33 °C to 35 °C) in ca 15 minutes. The actual brain temperature was monitored by fiber-optic probes (Luxtron, CA) (Fig. 1a). Measurements were performed on a 1.5 T MR scanner (Philips). The "patch" sequence was based on a standard 2D rf spoiled gradient echo [4, 5]. The spectral information was encoded by incrementing the echo time $TE_m = TE_1 + m\Delta TE$ (TE_1 = 6 ms) of the subsequent eight image records (m = 0, 1…7). Image matrix (256, 256), FOV = 200 mm and 128 phase encoding steps led to resolution in plane of 1.56x0.78 mm. The slice thickness was 5 mm, measurement time was 41 seconds (TR = 40 ms, 1 acquisition). Incrementing the echo time by ΔTE = 1.56 ms led to a spectral bandwidth 10 ppm. Spectral resolution was 1.25 ppm. To evaluate the accuracy of the proposed method, we performed experiments using a water phantom immersed in thermostatically controlled water bath. A set of six MRSI acquisitions in thermal equilibrium were performed at the temperature range 15°C - 40°C (Fig. 2). Reference temperature of the phantom and brain was measured before and after MRSI. The average of these temperatures was used for calibration. The processing algorithms were developed in house [4]. The measured matrix (k_{read}, k_{phase}, m) = (512,128,8) was zero-filled to a size (512,256,512). Data processing continued by 3D FFT. The resulting matrix (x_{read},y_{phase},δ) of the size (256,256,512), where x_{read}, y_{phase} are spatial coordinates and δ is the chemical shift, contained the proton magnitude spectra for each voxel (0.8x0.8x5 mm^3) of the measured slice. The voxel water spectral line position was determined from the centre of the linewidth at the half height [4].

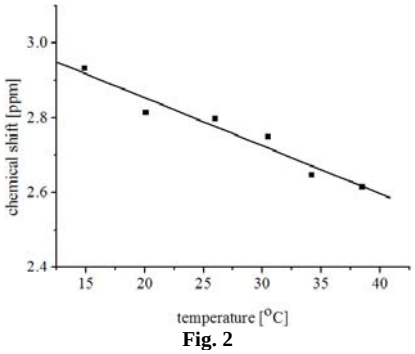

Fig. 2

Results: The relative temperature dependence of the water chemical shifts is shown in Fig. 2. The temperature coefficient of the water chemical shift (0.0123±0.001 ppm/°C) was determined from the slope of the regression line. In the typical cooling experiment the pig's brain temperature decreased by ~3°C during the first 20 minutes (Fig. 3). The mean temperature coefficient of the brain water chemical shift was -0.0192 ± 0.0019 ppm/°C (range from -0.0233 to -0.0164 ppm/°C). Figure 3 shows comparison of the temperatures measured by high-spatial-resolution SI (thin line) in the voxels V1 (Fig. 3a), V2 (Fig. 3b) and reference temperatures measured by fiber-optic probes (thick line). The voxels V1 and V2 (6 x 6 voxels, VOI = 0.1 cm^3) were placed at the ends of the fiber-optic probes (Fig.1). The relative temperature map in the pig's brain is shown in Fig. 1b. The temperatures reveal a selective decrease in the brain to be compared with the rest of the body. The cooling resulted in a temperature reduction at all sites, with somewhat larger fall (-4.7 °C) in the vicinity of the removed scalp.

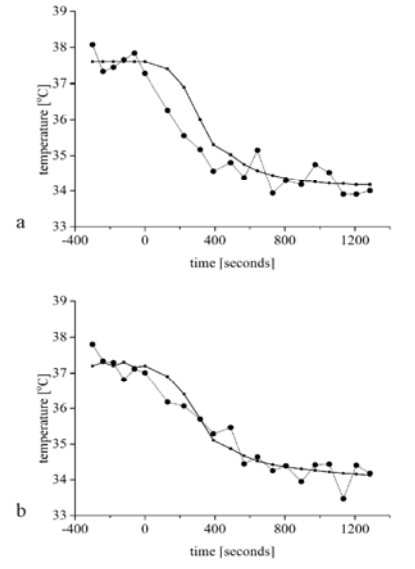

Fig. 3

Discussion: The temperature coefficient of the water chemical shift (-0.0123 ppm/°C) is somewhat more negative comparable with those from previous works (-0.01 ppm/°C) [6, 7]. In those experiments the temperature dependence of water chemical shift was evaluated relative to the reference spectral line (cyclopentane, DSS) whose chemical shift was assumed to be constant. There is however an important difference between the temperature coefficients when measured with and without an internal spectral reference. The microscopic local magnetic field B_{loc} also changes with temperature as a result of the temperature dependence of the magnetic susceptibility χ(T). The first-order approximation of B_{loc} can be expressed as [8,9] $B_{loc} = [1 - \sigma_{tot}(T)]B_{mac}$ where $\sigma_{tot}(T) = 2/3*\chi(T) + \sigma(T)$ is the total screening constant, σ(T) is the isotropic screening constant, and B_{mac} represents the macroscopic magnetic field. The temperature coefficient measured without an internal spectral reference is determined by the change of $\sigma_{tot}(T)$ with the temperature $\Delta\sigma_{tot}(T) = \sigma_{tot}(T) - \sigma_{tot}(T_{ref})$ where T_{ref} is the reference (baseline) temperature. $\Delta\sigma_{tot}(T)$ depends on $\Delta\sigma(T) = \sigma(T) - \sigma(T_{ref})$ and $\Delta\chi(T) = \chi(T) - \chi(T_{ref})$. On the other hand the temperature coefficient measured with an internal spectral reference is determined only by $\Delta\sigma(T)$. The influence of χ(T) is eliminated by subtraction of the water and reference spectral line positions. We consider our results in good agreement with the literature [6,7] data taking into account the susceptibility change of the water with temperature -0.0026 ppm/°C [8, 9]. The most important quantitative result of this work is the temperature coefficient of the brain water chemical shift -0.0192 ± 0.0019 ppm/°C measured without internal spectral reference. To our knowledge, there are no previous MRS measurements of this coefficient that can be compared with our value. Our coefficient could be explained by temperature dependence of the water magnetic susceptibility and by higher magnetic susceptibility of the biological tissues in lower temperatures. Note that the tissue type-dependent dynamic susceptibility effects caused by blood oxygen saturation, blood flow, perfusion, and susceptibility changes with temperature should also be considered in the living tissues.

Conclusions: Feasibility of temperature mapping with the proposed SI method has been demonstrated. The main advantage over the phase-mapping is avoiding the subtraction of subsequent measurements. Our results indicate that the proposed SI method can be used for monitoring the cerebral temperature changes during controlled hypothermia. Elucidation of the factors leading to more negative temperature coefficient of water chemical shift in the brain tissue measured without internal spectral reference compared to pure water needs further studies.

References: [1] Cady EB et al, Magn Reson Med 1995;33:862-867. [2] De Poorter J et al, Magn Reson Med 1995;33:74-81. [3] Covaciu L et al, Resuscitation 2008;76:83-88. [4] Weis J. et al, MAGMA 1997;5:201-212. [5] Weis J et al, Magn Reson Med 2007;57:22-28. [6] Schneider WG et al, J Chem Phys 1958;28:601-607. [7] Samson RS et al, NMR Biomed 2006;19:560-565. [8] De Poorter J. Magn Reson Med 1995;34:359-367. [9] Rieke V et al, J Magn Reson Imag 2008;27:376-390.