

A novel application for MR Thermometry: Post Mortem Interval estimation in forensic medicine

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Introduction The exact time point of death (the post-mortem interval, PMI) is of great importance in the legal and criminological field. Currently, the nomogram of Henssge is used worldwide for PMI estimation [1]. Input for this nomogram are the rectal temperature, body weight and ambient temperature, in combination with a correction factor regarding clothing or coverage of the body. This method has two major limitations: the minimal PMI margins are ± 2.8 hour and the maximum PMI that can be determined is $80 (\pm 7.0)$ hours. Currently, the rectal temperature used as input for the nomogram is a single point measurement. It is reasonable to state that the PMI estimation would improve with knowledge on the spatiotemporal cooling pattern of a subject. It is therefore hypothesized that temperature data at multiple time points and locations in the body increases the accuracy of the PMI estimation and extends the period of time in which results can be obtained.

MRI offers the possibility for non-invasive temperature measurements in 2D or 3D. The currently most widely used characteristic in MR thermometry (MRT) is the temperature dependence of the water proton resonance frequency (PRF). The water PRF shifts to lower frequencies with increasing temperature. It varies approximately linearly with temperature by about -0.01 ppm/ $^{\circ}\text{C}$ for pure water. Absolute temperature mapping can be performed by exploiting the frequency difference Δf between the PRF of water and of a temperature independent reference component such as fat [2]. MR spectroscopy (MRS) techniques measure resonance frequencies and can therefore be used for absolute MRT. The low spatial and temporal resolution of MRS techniques is of no concern in post-mortem imaging. Per temperature scan there are hardly any time constraints, since there is no subject motion. The goal of this work was to assess the feasibility of MRS based MRT to map absolute temperatures in post-mortem situations.

Materials & Methods An *ex vivo* bovine bone marrow (BM) sample was used for the experiments. Both water and fat are present in bone marrow, allowing for detection of both resonance frequencies and thus Δf_{wf} . The sample was cooled prior to scanning to a temperature of $T_{BM} = 7$ $^{\circ}\text{C}$. The cooled sample was placed uncovered on the MR table in the scanning room which was at $T_{room} = 22$ $^{\circ}\text{C}$. The sample temperature therefore increased over time during scanning. The room- and BM temperature were both continuously monitored by fiber-optic temperature probes (Luxtron, Santa Clara, CA).

Acquisition All scans were performed on a 3-T whole body system (Achieva, Philips, Best, The Netherlands). Eight scans were made, using a 2D MRS imaging sequence. In the center of a FOV (64×64 mm²), the signal from a volume of interest (VOI = $12 \times 12 \times 12$ mm³) was excited using PRESS. The number of phase encoding profiles was 16 in both directions. A total of 128 samples was acquired at a bandwidth of 1000 Hz. The echo time was $TE = 28.3$ msec, and repetition time $TR = 291.7$ msec. The number of averages was 4. Scan duration was 5 minutes.

Post-processing All post-processing was performed in Matlab (Mathworks, Natick, MA). For every scan, the average complex signal within the VOI was Fourier transformed. The water and fat frequency were determined separately from the absolute Fourier spectrum using a Lorentzian fit. The frequency difference Δf_{wf} was determined by subtraction and expressed in ppm. For each scan, the bone marrow temperature was determined by averaging the optic fiber measurements over the scan duration.

Results Figure 1 shows the absolute Fourier spectra of the complex MRS signal of all scans. In all spectra, both the water and fat peak are clearly visible. The bone marrow temperature is indicated per spectrum. The spectra show a shift of the water peak towards the fat peak with increasing temperature. Also the amplitude ratio between the water and fat peak changes with temperature. This may be caused by relaxation effects due to the relatively short TR used. In figure 2, the frequency difference Δf_{wf} is plotted against the bone marrow temperature. Clearly visible is the decreasing Δf_{wf} with increasing temperature. A linear fit through the data points yielded a temperature dependence of Δf_{wf} of -0.015 ppm/ $^{\circ}\text{C}$.

Discussion & Conclusion Our experiments have shown the feasibility of using MRS for the detection of water and fat frequencies, as well as their temperature dependent frequency difference, in *ex vivo* bone marrow. There is a clear correlation between Δf_{wf} and temperature. This is a good indication for the possibility of using MRS as an absolute temperature measurement technique based on bone marrow spectra. The temperature dependence of Δf_{wf} was found to be -0.015 ppm/ $^{\circ}\text{C}$. This value is very close to the reported value of -0.01 ppm/ $^{\circ}\text{C}$ for pure water. It should be noted that temperature might not be the only factor influencing Δf_{wf} [2]. All underlying processes affecting Δf_{wf} have to be studied extensively prior to the application of this technique. This is the focus of current work, in combination with analysis of both spatial and temporal temperature changes in post-mortem situations.

References [1] Henssge *et al.* Int J Legal Med. 2000 Oct;113(6):303-319 [2] Kuroda K. Int J Hyperthermia. 2005 Sep;21(6):547-60

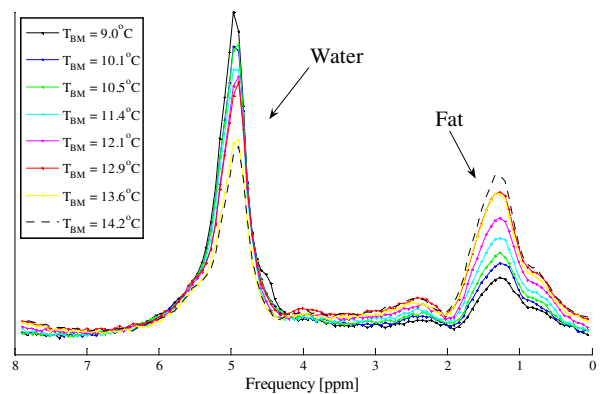


Figure 1 Absolute Fourier spectra of the complex MRS signal of all scans. The fat and water peaks are indicated. The bone marrow temperature is given per spectrum.

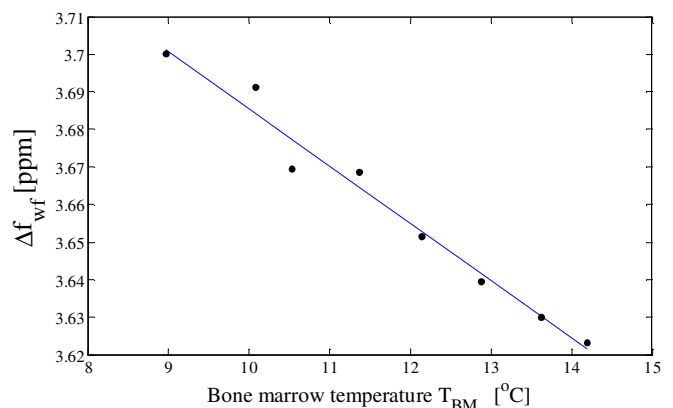


Figure 2 The temperature dependence of the frequency difference between water and fat. A linear fit is depicted with the blue line.