

POSSIBILITIES OF PHYSIOLOGICAL TEMPERATURE MEASUREMENT WITH MAGNETIC RESONANCE IMAGING

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INTRODUCTION

Non-invasive temperature mapping of the human body is a desirable technique for many problems in physiology, clinical medicine and ergonomics, such as the question of selective brain cooling, the application of hyperthermia and the verification of thermal models. For this purpose the use of magnetic resonance imaging (MRI) has been proposed.

MRI is based on electromagnetic signals which are evoked by high frequency pulses applied to the body causing the nuclear spin of the protons to precess with a certain resonance frequency, when placed in an homogeneous magnetic field, and then to relax again. The signal depends on the following three complex parameters defined in magnetic resonance imaging [1]: the proton density, the longitudinal (spin-lattice) relaxation time (T_1) and the transverse (spin-spin) relaxation time (T_2). Different tissues are mainly characterized by different values of T_1 and T_2 . Specific pulse sequences acquire images with certain contrast properties. T_1 -weighted images can be acquired by short repetition times (TR) between the pulses whereas long echo times (TE) between pulse and signal acquisition result in T_2 -weighted images. Mainly T_1 and the proton resonance frequency vary with temperature, and both parameters have been used to measure temperature in living human muscle [2, 3]. The molecular diffusion coefficient, which is also temperature dependent, can be measured by specialized sequences [4]. A disadvantage of methods using these parameters is the necessary post-processing procedure, which includes subtraction of images causing problems due to bulk tissue movements.

Another temperature dependent parameter is the equilibrium magnetization (M_0), equivalent to the spin density or proton density. In our study, the use of proton density weighted MRI sequences for measuring temperature changes has been evaluated both *in vitro* and *in vivo*.

MATERIALS AND METHODS

All MR measurements were performed with a 1.5 T system (Magnetom 63 SP MR, Siemens). The probes were positioned within the head coil of the MR system. The same attenuation was used for all imaging procedures of each experiment. For reference temperature measurements a Luxtron Fluoroptic Thermometer Model 3100

with a four sensor probe (type SMM) was used. This fiberoptic system allows temperature measurement during MRI procedure without any mutual interference.

In vitro experiments:

A piece of bovine muscle tissue (diameter approx. 10 cm) was first homogeneously warmed in a water bath. It was then cooled from the surface using a system of small parallel tubes with a continuous flow of temperate water. Four temperature sensor probes were implanted into the tissue at varying depths. During the cooling procedure, images were acquired every 5 min (spin-echo-sequence, TR = 900 ms, TE = 15 ms or 40 ms, FOV 120 mm, 256*256, slice 10 mm). The signal intensities of four regions of interest (5.5 mm²), one near the tip of each sensor, were evaluated.

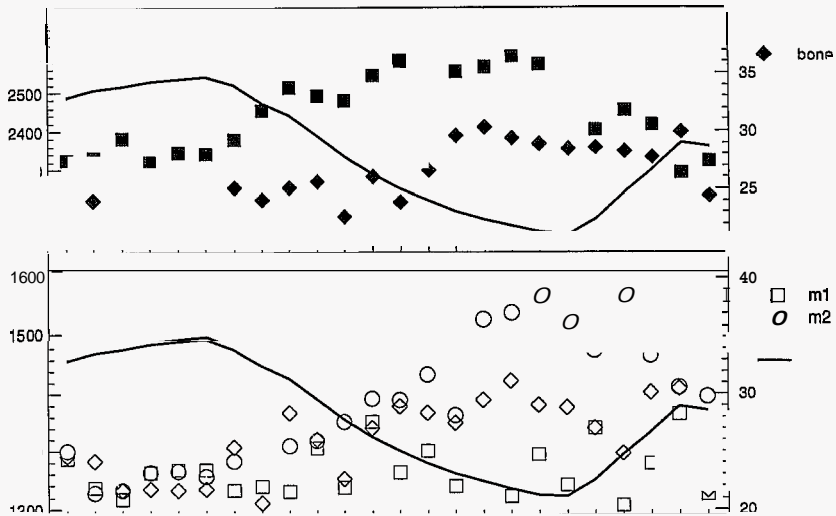
In vivo experiments:

The *in vivo* experiments were performed on four volunteers (2 females, and 2 males). The water perfused tubing system was wrapped around the right calf, covering a skin area of about 350 cm², corresponding to a leg length of 10 cm. The skin temperature was set by changing the temperature of the water pumped through the tubes. Skin temperature T_{skin} under the tube system was measured using a single sensor probe attached to the skin and isolated against the tubes. Images (TR = 1200 ms, TE = 10 ms, FOV 150 mm or 180 mm, 256*256, slice thickness 10 mm, acquisition time 5:11, orientation transverse) were continuously taken.

RESULTS

In the *in vitro* experiments, signal intensity of the muscle sample was higher at lower reference temperatures (T_{Ref}). The signal intensity was highly correlated ($r > 0.95$) with the reciprocal value of the absolute temperature ($1/(T_{\text{Ref}} + 273.15)$). Different locations of the sample at the same temperature had approximately the same signal intensity. The best accuracy obtained was ± 0.7 °C.

Fig. 1 shows the signal intensity of selected points within the leg and the skin temperature during an *in vivo* experiment. During cooling the signal intensity in the peripheral muscle tissue rose (m2 and m3), rewarming resulted in a lower signal intensity. No change was observed in the most central point (m1). Regarding the relationship of the signal intensity of the subcutaneous fat or the tibial bone marrow and the skin temperature, the signal intensity was also clearly higher at lower skin temperatures. These changes were observed in all *in vivo* experiments. The images had a good spatial resolution. The signal to noise ratio (tissue/background signal) was 37: 1.



CONCLUSIONS

Both the *in vivo* and the *in vitro* experiments demonstrated that the signal intensity of musculature generated by the sequences which were used for this study is higher for lower tissue temperatures. The sequences used in this study can be considered as proton-density-weighted, as the long TR minimize T_1 -weighting, very short TE minimize T_2 -weighting. The proton density as measured by MRI corresponds to the thermal equilibrium magnetization M_0 , which is inversely related to the temperature. This means that the equilibrium magnetization and therefore the measured proton density is expected to be lower for higher temperatures.

The signal intensities at the same tissue temperatures vary in the different experiments, depending on imaging parameters and the amplification of the MRI scanner. This variation makes it difficult if not impossible to obtain calibration curves *in vitro*, which could then be transferred to *in vivo* measurements.

In the *in vivo* experiments, there was a larger rise of signal intensity in the more peripheral muscle tissue during cooling. Assuming that the larger change of signal intensity corresponds to a greater fall of tissue temperature, this means that cooling the skin has a greater influence on the peripheral compared to the more central

muscle tissue temperature. No statement can be made regarding the magnitude of temperature changes or the absolute temperature within the tissue without invasive measurements. However, to localize a temperature change or to compare temperature changes in a homogeneous muscle tissue invasive measurements are not necessary. A comparison of temperature changes in different tissue types is not possible.

The type of sequences used in this study produces images which visualize temperature differences, at least in the muscle. Post-processing is simplified to a large extent. No calculations involving multiple images are necessary. Therefore, the presented method is less sensitive to problems with motion than other methods such as use of diffusion coefficient based techniques. The only artefacts observed here are motion artefacts caused by blood flow and the chemical shift at the subcutaneous fat, which do not hamper the evaluation. The signal to noise ratio is relatively high. The main disadvantage of this type of sequence is the acquisition time. This might be overcome by faster imaging techniques. Measuring tissue temperature with proton density weighted sequences is a promising and robust method, but more *in vitro* and *in vivo* studies are necessary.

However, MRI temperature measurement in general has some disadvantages. During the imaging procedure, the subjects must be placed in a narrow tube, where no metallic objects are allowed because of the strong magnetic field, and they must not move. These conditions restrict the general use of this technique for physiological purposes.

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