Non-invasive MR temperature imaging for temperature control of metallic implants

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INTRODUCTION:

In the past decade the safety of MR examinations in patients with metallic implants has been a constant object of investigation [1-5]. Such studies were performed in phantoms and often concentrated on temperature rise in the area surrounding the implants. Mostly the temperature was meassured with fiberoptic thermometers. Disadvantages of those thermometers result from their low spatial resolution (depending on their number of channels) and the fact that they can not be applied to in vivo studies or in vivo patient control in clinical routine. As a consequence the risk for patients can only be estimated in a global way, but not individually for each patient.

Our aim was to evaluate the feasibility of non-invasive MR temperature imaging in the vicinity of metallic implants for temperature monitoring. In our approach we used the proton resonance frequency (PRF) method that is based on the temperature dependence of the screening effect of bounded electrons and of the volume susceptibility constant [6]. This results in a shift of water PRF which can be evaluated from phase difference images.

MATERIAL AND METHODS:

The implantable device that was used for the experiments was a 4-pole intraspinal neurostimulation electrode made of titanium (Medtronic, Minneapolis MN, USA). It was placed in a phantom filled with agarose gel (1,5% agar in 0,5% NaCl solution). Three fiberoptic temperature sensors (Luxtron 790, Santa Clara, CA, USA; four channels, 0.1°C accuracy, 1s temporal resolution), used as a gold standard, were placed close to the electrode; one in direct contact to pole 3, two in the surrounding area in a distance of about 7mm from the electrode (Fig. 1). The fourth electrode was positioned outside the phantom as a reference. The spinal electrode was connected to a pulse generator for neurostimulation via a connection cable of 11m length positioned along the scanner's RF coils. In this setup heating of the electrode is ensured [5].

The studies were performed on a Magnetom Vision (Siemens, Erlangen, Germany) 1.5T whole body scanner with a flexible aray coil wrapped around the phantom. The phase images were acquired with a standard MRI FLASH sequence without phase correction (TR = 28ms, TE = 15ms, flip angle = 30°, FoV = (230mm)², matrix = 256²) within 9s per image and a nominal resolution of 0.90x0.90 mm^2. The reference phase image (normal temperature) was averaged over 10 acquisitions. Before imaging the automatically performed shim was manually improved to minimize deletions near the metallic objects.

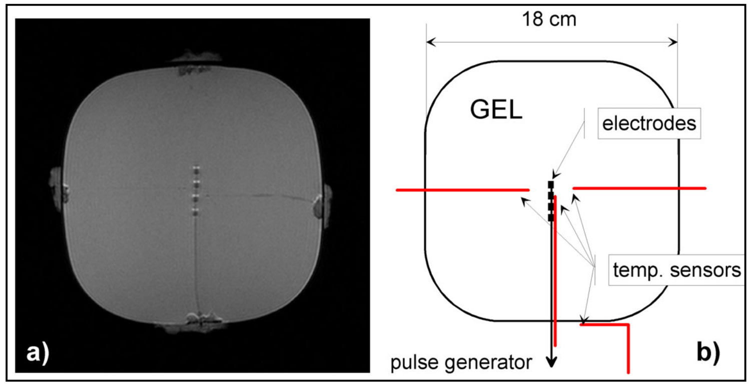

Fig. 1: Experimental setup; a) high resolution coronal TSE image; b) corresponding schematic illustration

The electrode and its surrounding gel was heated by a high SAR T1-spin echo scan for 10 min. In this time temperature at the electrode rose from 24.9°C to 61.0°C, in the surrounding gel from 25.7°C to 42.2°C and from 25.2°C to 42.9°C. Directly after the SE sequence four phase images at high temperature were acquired (9s, 18s, 27s, and 36s after the SE sequence). 15min. after heating the reference phase image was done (Fig. 2). From the phase difference images temperature difference maps were calculated (Fig. 3).

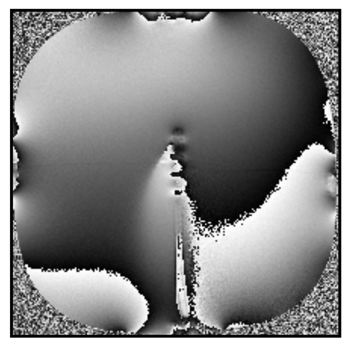

Fig. 2: Reference phase image 15 min. after heating

RESULTS:

Fig. 3 shows the color encoded temperature map based on the subtraction of the first phase image and the reference scan 15 min. after the SE sequence. The resolution of the temperature map is good enough to resolve the structure of the electrode: The temperature of the electrode tip with its four metallic contacts is higher than the connecting cable which is coded with a layer of polyurethan. The dispersion of heat in the gel surrounding the electrode is clearly visible. The calculated values correspond very well to the values measured by the fiberoptic thermometer. After heating the temperature time course was monitored by four phase images. There was no significant differences between the four images showing that the cooling process was too slow to result in measurable temperature differences within these first 36 seconds.

Deletions of the MR signal at the electrode due to transitions in susceptibility were visible. The were too small, however, to be a restriction of the method.

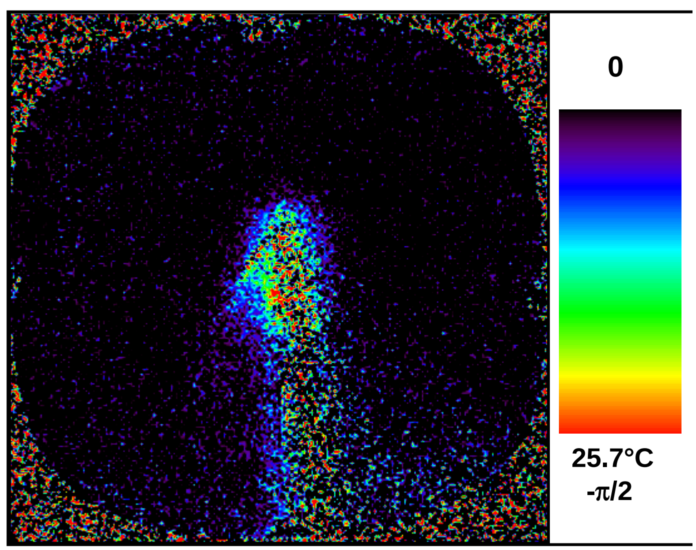

Fig. 3: Color encoded temperature difference / phase shift

DISCUSSION:

Non-invasive MR temperature imaging based on proton resonance frequency shift has been used for control in hyperthermia applications [6]. In this study it could be shown that it is a suitable method for temperature control during MRI in patients with metallic implants, as well. The temperature map that was acquired combines high spacial with high thermal resolution (up to 0.2°C in phantom studies [6]) which ensures that even small and spacially limited heating can be registered. With an acquisition time of 9s per image this method also registers fast and spontaneous heating which is reported in the literature [5].

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