Real Time Measurement of Heating Near Metallic Implants Throughout a Phantom Using Phase-Shift MR Thermometry

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Introduction – Potentially dangerous heating occurs at the narrow tip of certain medical implants in patients being imaged by standard MRI systems [1, 2]. This heating is induced by RF fields (B1 field). Fiber optic thermometers are specified in ASTM F2182 [3] for measuring this type of MRI-induced local heating at a single or a few points near medical implants. F2182 specifies 1 sample every 12 s. Finding small (< 8 mm^3) local hot spots in an entire phantom with these probes is difficult. Previous attempts to measure heating near a spinal cord stimulator electrode (SCSE) via phase shift thermometry required 10 minutes of MRI heating before the temperature was measured. Georgi reported a 36.1 °C temperature rise at the electrode and a 16.5-17.7 °C in the surrounding gel [4]. This long heating time and high temperature rise caused large thermal diffusion errors and did not comply with the ASTM 12 s sampling rate. We explored imaging of heating in real time via phase shift thermometry in phantoms using a clinical MRI system [5]. Our goal was to monitor, every 3-5 seconds (s), the temperature rise inside a phantom containing an insulated wire with a bare metal end and other medical implants. These MRI-exposed objects induce very high local (point) SAR and heating at their exposed metal ends.

Material and Methods – We used the following components: *1. <u>Phantom</u>* – a plastic box 40 cm x 27.4 cm x 14.6cm with wall thickness of 1.5 mm. The phantom was filled with 9.5 liters of gel (0.55% saline, 1% hydroxyethyl cellulose gelling agent, 1.0 % $CuSO_4$). *2. <u>Implant</u>* – an insulated copper wire 79.5 cm long with a diameter of 1.5 mm and one end with insulation removed for 1cm. The bare tip had a diameter of 0.6 mm. This wire was suspended 5.8 cm above the bottom of the phantom with a plastic grid and was looped so that its two ends were separated by 6 cm. The wire was not connected to anything *3. <u>MRI System</u>*- We used a 3.0 T Intera system (Philips Medical Systems, NL) with a sense torso coil. Near real-time MR thermometry was performed with a 2D-multislice T1 weighted fast field echo (T1W FFE) sequence [5]. We experimented with various sequence parameter settings. For wire heating and thermometry, a representative sequence had field of view 300 x 300 mm, rectangular field of view 70%, 3 slices of 3 mm thickness and 0.3 mm spacing, TR 53 ms, TE 20 ms, flip angle 30°, signal average 1, sense 3, and 9.2 s per 3 slices. The acquisition spatial resolution of the image was 1.25 x 1.55 x 3 mm; the reconstructed spatial resolution was 1.17 x 1.17 x 3 mm. Phase images were reconstructed in real time and analyzed on a PC with our custom software that performed phase subtraction, phase unwrapping, and temperature plots. Simultaneous thermometry was performed during imaging with a fiber optic temperate probe (Luxtron, Santa Clara, CA) at selected points in the phantom and at the wire tip. The above sequence had a system-computed whole body (WB) SAR of 12.4 W/kg. This SAR is not necessarily equal to the measured SAR. In separate experiments with a similar sequence the system indicated a WB SAR 4.7 times higher than we measured in our phantom with calorimetry.

Results – We obtained phase images of the temperature distribution in 3 slices every 9.2 s. Once the plane with the center of the wire tip was located, we imaged this plane every 3.1 s for a period of 108.5 s or longer (fig. 1). Temperatures versus time for the wire at points near the tip were processed and plotted. In addition to the MRI-determined temperature, we recorded the temperature versus time at the tip of the wire with a fiber optic thermometer (fig. 2). Locating the maximum heating near tips of thin wires can be tedious since the area of the initial temperature rise is very small (< 1 mm).

Discussion - We demonstrated real time (3.1 s per image) acquisition of MRI-induced heating near the tip of a simulated metallic medical device lead (an insulated a wire with exposed tip). Implants with small (< 2 mm) bare metallic tips at the end of long insulated wires represent a worst-case heating in intense RF fields. This preliminary MRI thermometry provided a resolution of better than ± 0.5 ^{0}C, reconstructed spatial resolution of 1.17 x 1.17 x 3 mm, and a sample rate of 3.1 sec. These are adequate for determining heating per ASTM F2182 near even the smallest implants. This method has the potential to be a valuable tool for viewing and measuring MRI-induced heating near implants in real time. This includes heating of a phantom or a body in vitro and in vivo. Non-heating image artifacts from fiber optic thermometers and wire should be able to be removed by background subtraction since they do not change over time. One requirement for small phantoms is the need to increase (via software patch) the WB SAR significantly above the SAR limits of the MRI machine set by the manufacturer. We are performing studies of actual implants such as spinal cord stimulator electrodes. In preliminary measurements we observed large image artifacts surrounding the proximal end of a large metallic connector of a spinal cord stimulator electrode, and to a much smaller extent at the distal tip. This may limit the utility of MRI thermometry for viewing heating of certain implants.

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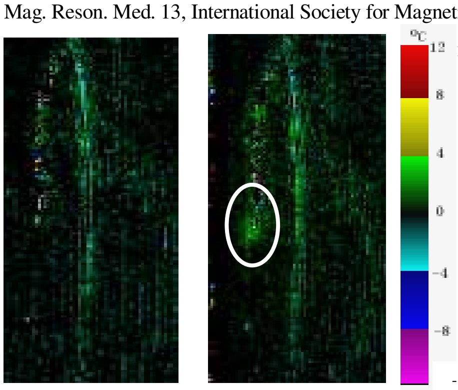

Figure 1. Initial (left) and final (right) temperature maps in the plane of the wire tip. Circle indicates tip heating. Other lines are non-heating artifacts from fiber optic thermometer and wire.

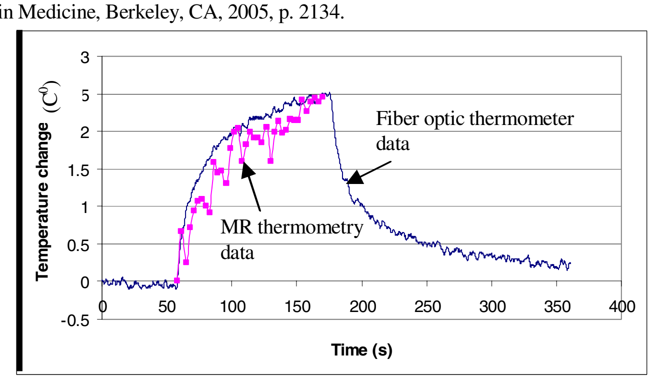

Figure 2. Temperature elevations versus time near the wire tip. MR thermometry data is averaged over 3 x 3 pixels.