Evaluation of RF and resistive heating of magnetically- assisted remote control (MARC) steering coils in endovascular interventional MRI

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INTRODUCTION:

Using 1.5 F catheters in a 2 T scanner, Roberts et al. (1) demonstrated that current applied to orthogonally arranged solenoid and Helmholtz coils wound at the tip of a catheter can be used to steer a catheter tip through simulated vessel branch points. The magnetic moment, M, created in a each coil, is given by the equation,

$$M = nIA,$$

where n, is the number of turns in the coil, I, is the current applied, and A, is the cross-sectional area of the coil. The net magnetic moment vector produced by the coils experiences a torque $M_{(i, j, k)} \times B_{(k)}$ which deflects the distal catheter tip in the main magnetic field, B. For a given deflection, the magnetic torque necessary must overcome the "stiffness" of the catheter which is related to its Young's modulus, moment of inertia, and unconstrained length (2). This requires currents of up to 200-300 mA (1, 2) The forces generated by this method offer a unique opportunity for steering endovascular catheters in an endovascular interventional MRI setting. Nonferromagnetic conducting metals used in guidewires, braided catheters, active catheter tracking coils, or intravascular imaging coils, however, have been shown to cause significant heating as a result of radiofrequency (RF) heating due to the antenna effect (3, 4). The purpose of this study was to evaluate the temperature increases produced by RF heating of the steering coils during real-time MR imaging, and the temperature increases resulting from resistive heating after application of currents necessary for deflection.

MATERIALS AND METHODS:

A three-axis coil using 44 AWG magnet wire (California Fine Wire, Grover Beach, CA) was wound on the tip of a 1.8 F Baltacci catheter (BALT, Montmorency, France). To assess a worst-case heating scenario without blood flow, in vitro testing was performed in a 1.5-T/64-MHz MR system (GE Medical Systems, Wakesha, WI) using a tissue –mimicking agarose gel-filled phantom (2% agar, 0.9% NaCl) designed to approximate an artery with the same thermal properties of blood (thermal conductivity = 0.6–1 S/m, relative permittivity = 73.6, specific heat = 4.2 kJ/kg.K). The phantom set up was aligned parallel to the long axis of the magnet bore. The temperature was recorded with a fiberoptic temperature system (model 3204; Luxtron Corp., Northwestern Parkway, CA) onto a PC via a serial cable using an RS232 protocol and Microsoft HyperTerminal. Four Luxtron temperature probes were fixed within the gel at different locations in close proximity to the catheter tip coils. RF heating of the coils was measured at 0, 5, 10, and 15 cm from the magnet isocenter during 15 min of continuous true-FISP (TR= 4.2 ms, TE=1.6 ms, α =75º, slice thickness = 5 mm, matrix = 128 x 128, FOV = 20 cm, SAR = 4.3 W/kg), and with varying flip angles between α = 5º and 90º, at a distance 15 cm from iso-center. Resistive heating was measured during application of DC current from 50-300 mA for 30 seconds to each coil individually and in combination using the same true-FISP pulse sequence, off-center position, and fiberoptic probe configuration.

RESULTS:

In the absence of applied current, higher RF heating-induced temperature changes were observed at higher distances from the magnet iso-center. A 0.35 ºC increase was measured at 15 cm from isocenter whereas no increase in temperature was seen at magnet isocenter after 15 min of real-time imaging (Figure 1). Higher RF flip angles also resulted in increased temperature, for example a 0.25ºC increase was observed using 80º flip angle, whereas a 25º flip angle resulted in a 0.02ºC increase (Figure 2). Larger temperature increases were seen after application of current to the coils (Figure 3). After 30 seconds of application of 300 mA to one coil, a 2ºC increase was observed. After 30 seconds of application of 300 mA to three coils simultaneously, a 9 ºC increase was observed.

CONCLUSION:

The RF induced temperature increases on this MARC-steered catheter are minimal and dependent on the distance from the magnet bore wall and the power of the pulse sequence. Physiologically significant resistive heating (greater than 4 ºC) occurs only with currents above 200 mA in more than one coil. Currents below 200 mA in one or more coils produce relatively minor, physiologically inconsequential temperature increases. In practice, convective blood flow, more efficient catheter coil systems designed to maximize deflection per unit of current applied, and catheter materials with higher specific heat capacities could be used to decrease resistive heating. MARC steering may be an attractive alternative to traditional endovascular catheter navigation with heating safety concerns arising only at higher currents using this prototype.

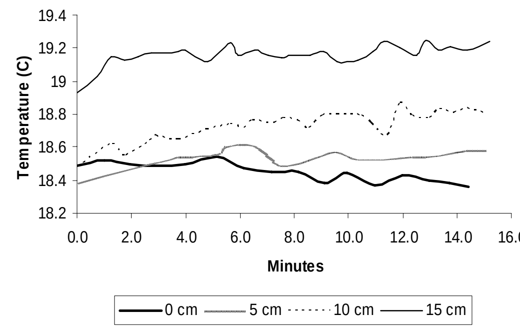

Figure 1. RF heating-induced temperature changes after 15 min of true-FISP imaging with increasing distance from the magnet iso-center.

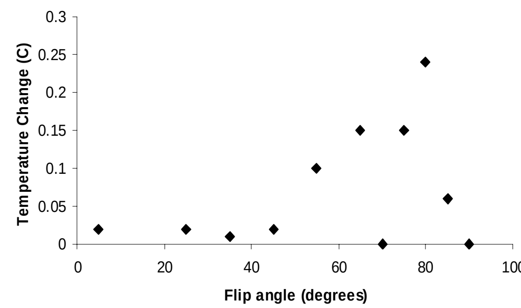

Figure 2. RF heating-induced temperature changes as a function of True-FISP flip angle.

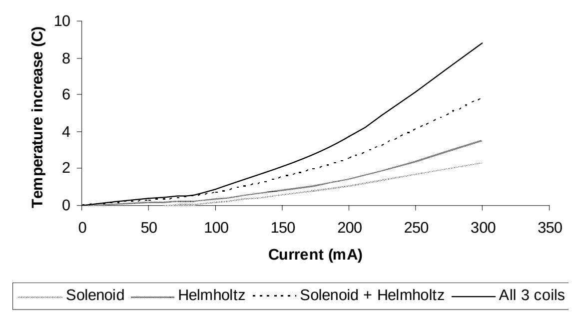

Figure 3. Resistive heating-induced temperature changes with current applied for 30 seconds to solenoid and helmholtz coils, individually, and in combination.

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