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RF HEATING OF MRI-ASSISTED CATHETER STEERING COILS FOR INTERVENTIONAL MRI

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Abstract

RATIONALE AND OBJECTIVES—To assess magnetic resonance imaging (MRI) radiofrequency (RF) related heating of conductive wire coils used in magnetically steerable endovascular catheters.

MATERIALS AND METHODS—A 3-axis microcoil was fabricated onto a 1.8 Fr catheter tip. *In vitro* testing was performed in a 1.5 T MRI system using an agarose gel filled vessel phantom, a transmit/receive body RF coil and a steady state free precession (SSFP) pulse sequence, and a fluoroptic thermometry system. Temperature was measured without simulated blood flow at varying distances from magnet isocenter and varying flip angles. Additional experiments were performed with laser-lithographed single-axis microcoil-tipped microcatheters in air and in a saline bath with varied grounding of the microcoil wires. Preliminary *in vivo* evaluation of RF heating was performed in pigs at 1.5 T with coil-tipped catheters in various positions in the common carotid arteries with SSFP pulse sequence on and off, and under physiologic flow and zero flow conditions.

RESULTS—In tissue-mimicking agarose gel, RF heating resulted in a maximal temperature increase of 0.35°C after 15 minutes of imaging, 15 cm from magnet isocenter. For a single axis microcoil, maximal temperature increases were 0.73-1.91°C in air and 0.45-0.55°C in saline. *In vivo*, delayed contrast enhanced MRI revealed no evidence of vascular injury and histopathological sections from the common carotid arteries confirmed the lack of vascular damage.

CONCLUSIONS—Microcatheter tip microcoils for endovascular catheter steering in MRI experience minimal RF heating under the conditions tested. These data provide the basis for further *in vivo* testing of this promising technology for endovascular interventional MRI.

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Keywords

Catheter; interventional; MRI; heating

INTRODUCTION

Current applied to copper microcoils **placed** at the tip of an endovascular device can be used to steer a catheter within a clinical magnetic resonance imaging (MRI) scanner [1,2]. The current applied induces a magnetic moment within the coils at the catheter tip, which experiences a torque in the presence of the scanner main magnetic field, producing remotely controllable deflections that can be used for navigating vessel turns and branches or holding the catheter tip in a specific orientation. The electric currents required to generate the desired magnetic moments, however, can generate enough heat through resistive dissipation to cause temperatures unsafe for blood or vascular walls. We have shown that undesirable temperature increases due to resistive heating can be mitigated using high thermal conductivity material for the catheter tip and flowing saline coolant through the catheter lumen (please see accompanying manuscript "Steerable Catheter Microcoils for Interventional MRI: Reducing Resistive Heating").

Undesirable heating, however, may also occur in a long wire (e.g., a guide wire) running down the lumen of the catheter due to field coupling with radiofrequency (RF) magnetic fields used for MR imaging through the "antenna effect" [3,4]. Metallic devices used in MRI-guided endovascular procedures, such as guidewires, braided catheters, catheter tracking coils, or intravascular imaging coils, even if non-ferromagnetic, have the potential to interact with the time-varying RF electromagnetic fields from the RF transmitter, causing heating in and near the device, and thus posing a hazard to the patient The concentration of RF power within such devices may cause a significant increase in the specific absorption rate (SAR) of the surrounding tissue, potentially resulting in excessive local temperature rise and burns [5-7].

Beyond the well-known case of metallic implants that can cause heating hazards [8], several authors have investigated the specific case of metallic wires in the setting of interventional MRI [9-12]. Most studies have shown that the specific absorption rate measured in the presence of a metallic wire can exceed the SAR limitation of 2 **Watts**/kg [5-7]. The FDA-recommended temperature rise due to SAR should not exceed 1°C on or in the head and 2°C in the torso and extremities [13]. Safety considerations are, therefore, integral to the development of these promising techniques.

A number of factors have been shown to determine the amount of RF heating due to the antenna effect, including wire geometry (diameter and length), distance from isocenter (since coupling occurs between to the electric field of the transmit coil and the wire), and power of the RF pulses [5-8,14,15]. These studies, however, are concerned with single wires whose distal end is an open circuit. In the case of microcoils used for catheter tip deflection, current carrying wires transit the length of the catheter lumen in pairs. Each pair is connected at the distal end to a conducting microcoil with low inductance. In the case of direct current, each individual microwire carries equal and opposite current with respect to the other microwire of the pair. Magnetic fields from the pair tend to cancel each other out except between the wires. This is also true for alternating currents for which there is not a significant phase shift (> 90°) across the wire pair. Considered as a receiver, therefore, a field between the wires of a pair (and perpendicular to the plane of the wire pair) is required to excite a current in the wire and fields elsewhere are ineffectual. Thus, the coupling of a wire pair with a field would be expected to be weak for the wire pair compared with the single wire dipole antenna at RF frequencies used in clinical MR pulse sequences.

In the first set of experiments in this study, we measured temperature changes related to RF heating of hand-**made** microcoils in a tissue-mimicking agarose gel-filled vessel phantom using MRI compatible thermometric fiberoptic probes. Experiments were performed in a 1.5 T/64 MHz MR system using the transmit/receive body RF coil and a steady state free precession (SSFP) pulse sequence. This pulse sequence has been previously shown to have optimal spatial and temporal resolution for catheter-coil tip visualization. Temperature changes of the coils and surrounding gel were measured under a variety of circumstances known to affect RF heating, including varying distance from the magnet isocenter and varying flip angles.

In the second set of experiments, also in a 1.5 T MR system using SSFP imaging sequence, temperature changes related to RF heating of laser-lithographed microcoils were tested in air and in a saline bath in the lateral aspect of the MR bore in order to maximize potential RF heating.

The third set of experiments consisted of *in vivo* testing of microcoil-catheter constructs in a combined X-ray angiography and MRI (XMR) suite. A single axis solenoidal alumina-tipped catheter with luminal saline drip was navigated under x-ray guidance to the common carotid artery (CCA) of a pig via percutaneous transfemoral access. The pig was subsequently moved to a 1.5 T clinical MR scanner via floating table. The catheter was advanced at 2 cm increments in the CCA with SSFP imaging increasing from 0 seconds to 5 minutes of activation with each 2 cm catheter advancement. The catheter was then navigated to the contralateral CCA, the CCA was ligated proximal to the catheter tip (in order to achieve zero flow conditions in the artery) and the experiments were repeated. Delayed contrast enhanced MRI was performed after the administration of 0.15 mmol/kg Gd-DTPA to evaluate potential vascular damage. The animals were euthanized after the imaging session and both common carotid arteries were retrieved for histopathologic assessment for thermal or mechanical damage to the vessel walls.

MATERIALS AND METHODS

Hand Made Coil Testing

Device construction—A 1.8 F Baltacci catheter (BALT, Montmorency, France) was obtained and the most distal catheter tip containing heavy metal marker was cut to eliminate confounding magnetic forces and MR artifact. A three-axis coil using 44 AWG magnet wire (California Fine Wire, Grover Beach, CA) was wound on the tip of the modified catheter. The z-axis coil was solenoidal and consisted of 100 turns. The two orthogonal modified Helmholtz coils were wound along the side of catheter over the solenoid coil and fixed with adhesive glue. The Helmholtz coils generate a magnetic moment perpendicular to the long axis of the catheter (perpendicular to the solenoid). Approximately 40 turns were used in each of the Helmholtz coils, but the effective area inside the coils was approximately the same as that of the solenoidal coil. The final outer diameter of the catheter tip with 3-axis coils was 4 mm. The length of the coil assembly was approximately 8 mm. Note that, given the cylindrical geometry of the catheter, the Helmholtz coils might be more readily described as paired "racetrack" windings, to recognize their elliptical form and the catheter-specific separation (whereas a true Helmholtz coil consists of a pair of coaxial circular windings of radius equal to separation). Loose wire ends were braided (to prevent local field inhomogeneity artifact from the wire) and wrapped around the catheter along its length.

Temperature measurements in the MR system—To assess possible RF heating of the solenoid and modified Helmholtz coils, temperature measurements were performed in a 1.5T MR system (Magnetom, Siemens Healthcare, Erlangen, Germany). All temperature measurements were conducted with the catheter embedded over a distance of 1 m in a tissue-mimicking gel (2% agar, 0.9% NaCl) that prevents thermal convection and has electrical properties similar to human blood (thermal conductivity = 0.6–1 S/m, relative permittivity = 73.6, specific heat = 4.2 kJ/kg.K). The gel and catheter were contained within a cylindrical plastic phantom of 5 cm in diameter. The phantom set up was aligned parallel to the long axis of the magnet bore. The temperature was measured with fiberoptic temperature probes (model 3204; Luxtron Corp., Northwestern Parkway, CA). The probes are non-metallic and immune to the radiofrequency magnetic fields of MRI. The length of the probes extended 10 m in order to extend outside of the MR scanner suite. Temperature values were recorded on a PC via a serial cable using an RS232 protocol and Microsoft HyperTerminal. Four Luxtron temperature probes were fixed within the gel, adjacent to the catheter tip using as follows (Figure 1): Probe 1, at the distal tip of the solenoid coil; Probe 2, between two paired Helmholtz coils; Probe 3, at the more proximal end of solenoid; Probe 4, 2 cm away from catheter-coil tip. During all experiments identical sensor positions are ensured by adhesive fixation. The tip of the sensor was not masked by the tape, so that the temperature increase of the adjacent gel could be measured. All experiments were carried out without any applied current to isolate temperature increases related to RF heating. The sampling rate was one data point per second.

To investigate the effect of position within the scanner bore, temperature measurements were performed at off-center distances (x-position) between 0 cm (magnet center), 5 cm, 10 cm, and 15 cm (close to magnet bore). A SSFP pulse sequence (TR= 4.2 ms, TE=1.6 ms, α =75°, slice thickness = 5 mm, matrix = 128 × 128, FOV = 20 cm) was used, providing a specific absorption rate of SAR = 4.3 W/kg. At each off-center position the sequence was applied over a period of 15 minutes. To investigate the effect of pulse sequence power deposition during scanning, temperature measurements were also performed over a period of 10 minutes for different flip angles between 5° and 90°. A SSFP pulse sequence (TR= 4.2 ms, TE=1.6 ms, slice thickness = 5 mm, matrix = 128 × 128, FOV = 20 cm) was again used.

Laser Lithographed Coil Testing

Device construction—A single axis Helmholz coil (of approximately 10 nanoHenry inductance) was fabricated on an alumina tube and this construct was attached to a substrate microcatheter and conducting 36 gauge (0.127 mm) copper wires were drawn in parallel through the lumen of the 150 cm microcatheter as described in the accompanying manuscript, Materials and Methods, “Microcoil-Catheter Construction.”

Radiofrequency Heating Evaluation—To assess RF heating of microcoil-catheter assemblies, the coil-tipped catheters were attached to 1.4 mm diameter MR-compatible fluoroptic fiber optic temperature (FOT) probes (Luxtron, Santa Clara, CA, USA). The copper wires soldered to the coils were run along the length of the FOT probe, and the entire coil-wire-probe construct was sealed in heat shrinkable tubing. This served to insulate the coils and FOT probes from significant ambient heat loses and confined RF-mediated heating to the FOT probe. The coil-wire-probe constructs were maximally offset (27.4 cm) from the bore isocenter within in the lateral groove of a 1.5 Tesla, short-bore clinical MR imager (Achieva, Philips Healthcare, Best, The Netherlands) under several conditions as outlined in Table 1 and Figure 4.

The electrical circuit was designed to function with DC input. The circuit is dominated by the electrical properties (inductance, capacitance and resistance) of the lead wires. The microcoil inductance makes only a small additive contribution to the lead wire inductance and a larger but still fractional addition to the lead wire resistance. It is not expected to resonate in the 64 MHz field of the MRI although no steps were taken to mitigate this potential effect. For most experiments, the coil / FOT probe construct was suspended in air. For the final two experiments, the coil / FOT probe construct was suspended in a 2 cm diameter tubular saline-containing phantom (3 cm diameter, 100 cm length). Prior to imaging, the construct was manually moved through the length of the bore with the MR scanner RF coils inactive. Lead wires were attached to a telephone cable which was, in turn, run through the door of the scanner room into the control console area, where the phone jack adaptor, BNC connector, and power supply were located. For all experiments in this series, imaging was carried out with steady state free precession (TR = 5.5 ms, TE = 1.6 ms, flip angle = 30°, 128 × 128 matrix, 5-6 mm slice thickness). No current was applied to the microcoils in order to eliminate resistive heating. Each scenario was carried out for a period of 70 seconds, with 10 seconds of baseline temperature sampling, then 30 seconds temperature sampling during RF exposure (SSFP sequence active), followed by 30 seconds of further temperature sampling without RF exposure. The sampling rate was one data point per second.

Preliminary In Vivo Evaluation at 1.5 Tesla—Limited *in vivo* evaluation of the potential for RF heating of the coil-tipped catheter was performed in anesthetized porcine model. As described in detail in the accompanying manuscript on catheter resistive heating, a 150 cm long 2.7 F Tracker-18 microcatheter (Boston Scientific, Fremont, CA) served as the substrate for an alumina-tipped catheter with a hand-**made** 75 turn solenoid coil. Lead wires were attached to microcoil proximal to the catheter tip. The leads were pulled through a modified Thuoy-Borst Y-adaptor at the microcatheter hub and were subsequently attached to the catheter. Power leads were brought through the center bore of the Y-adaptor, allowing saline to be infused through the side port.

Under a protocol approved by the Institutional Animal Care and Use Committee, a 30 kg farm pig served as a test subject to investigate potential thermal damage from resistive and RF heating related to use of the microcoil-tipped catheter. Anesthesia was performed as outlined in the accompanying manuscript on resistive heating. This interventional study was performed in an XMR suite. A 9 F sheath was inserted into the femoral artery percutaneously followed by 50 IU/kg heparin. A 90 cm long 9 F guide catheter was navigated over a guidewire to the proximal portion of the left CCA under x-ray guidance. The coil-tipped microcatheter construct was then inserted through the guide catheter and advanced to the distal CCA, 12 cm beyond the guide catheter tip, or a total of 102 cm from the access site in the femoral artery of the pig. The animal was then transported to the MRI scanner, with the femoral access site just outside the opening of the magnet bore and the carotid arteries near the isocenter. Because UCSF has a combined X-ray/MR suite, animals can transition from the X-ray angiographic suite to the MR imaging suite on a single sliding table. Imaging and catheter heating experiments were then performed in the left CCA as outlined below.

In the x-ray angiography suite and the proximal right CCA was catheterized with the 9 F guide catheter. The coil-tipped microcatheter construct was then inserted through the guide catheter and advanced to the distal right CCA. The right CCA at the tip of the 9 F catheter but well proximal to the microcoil-tipped catheter tip was then ligated with a silk suture in order to obtain zero flow in the CCA. The animal was then transported to the MRI scanner. Imaging and catheter heating experiments were then performed in the left CCA as outlined below, again with initial insertion of the microcoil tipped catheter 12 cm beyond the guide catheter tip, or 102 cm from the femoral artery access site.

Testing was performed using a 1.5 Tesla clinical MR scanner. Microcatheters were tested at six positions in each CCA, with the most distal location being the first test point, and each subsequent test point separated by a 2 cm manual pull back of the catheter to ensure an adequate margin between test points in case thermal damage occurred at any given test point. As outlined in Table 1 of the accompanying manuscript, the initial 3 test points in each artery (102 cm, 100 cm, 98 cm) were directed primarily at resistive heating caused by running current (300 mA for 30 seconds, 1 minute, or 5 minutes) through the catheter microcoils during imaging and the subsequent 3 test points (96 cm, 94 cm, 92 cm) in each artery were directed primarily at RF heating in catheters receiving no current by just being imaged (0 mA for 30 seconds, 1 minute, or 5 minutes). Imaging was performed with a SSFP sequence (TR = 5.5 ms, TE = 1.6 ms, flip angle = 30°, 128 × 128 matrix, 3mm slice thickness, SAR = 3.7 Watts/kg), with real-time imaging at 3 - 5 frames per second. Left CCA experiments were at physiologic arterial flow; right CCA experiments were at arterial stasis by occluding the artery. For all experiments, room temperature normal saline solution was perfused through the central lumen of the microcatheter at a drip rate of approximately 0.1 ml/s.

Delayed contrast enhanced MRI (DE-MR) was performed to evaluate arterial wall damage. DE-MR images were acquired after injecting another 0.05mmol/kg Gd-DTPA. These images were obtained 5-10 minutes after injection. Images were acquired in axial- and coronal views encompassing the common carotid artery using an inversion recovery gradient echo sequence (TR/TE/flip = 5ms/2ms/15°, shot interval = 2RR-intervals, slice thickness = 3mm, no slice gap, FOV=26×26cm, matrix size=256×162).

At the conclusion of the vascular intervention and imaging, the guide catheter and microcoil-tipped catheter were removed. The animal was maintained under anesthesia for an additional two hours in order to allow time for histologic changes from potential thermal damage to evolve. The animal was then euthanized and histologic specimens were obtained, as outlined in the accompanying manuscript.

RESULTS

RF Heating of Hand Made Coils in Agarose Gel Phantom

In the absence of applied direct current, experiments were carried out at various distances from the magnet bore wall to determine RF heating effects induced in the solenoid and Helmholtz catheter coils due to RF pulse sequences during real-time MR imaging. At the isocenter of the magnet, no heating was detected by any of the probes after 15 minutes of imaging (Figure 2A). As the apparatus was moved 5 cm closer to the magnet bore wall, a slight increase in the temperature (0.2 °C) recorded over the 15 min period was measured in probe 1 and probe 2 (0.14 °C), close to the distal solenoid tip and Helmholtz coils (Figure 2B). Furthermore, the change in temperature in probes 1 as well as probe 2 increased as distance to the magnet bore wall decreased (Figure 2B-D). Probes 3 and 4 showed no significant change in temperature at all distances from the bore (Figure 2A-D). At any distance from isocenter, however, the increase in temperature over the 15 minute time interval was not higher than 0.35 °C.

The power of RF pulses increases with increasing flip angles. Higher RF pulse flip angles resulted in increased RF heating of the catheter-coil (Figure 3). The increase in temperature observed was small, however. After 10 min of continuous MR imaging 15 cm from isocenter, 0.25 °C was the largest increase in temperature observed (80° flip angle).

Radiofrequency Heating of Laser Lithographed Coils in Air and Saline

The results of the RF heating experiments are shown in Table 1 and Figures 4 and 5. The maximal increase in temperature above baseline was about 1.9°C, which occurred with the SSFP sequence active for 30 seconds for a catheter in air with leads grounded or floating (Figure 5A). The smallest temperature increase was about 0.5°C, which occurred in when the catheter was suspended in a saline filled tubular phantom for 30 seconds (Figure 5B).

In Vivo Results

Under the conditions of normal arterial flow, delayed contrast enhanced MRI revealed no evidence of vascular injury related to catheter navigation or heating. These findings were confirmed by gross and microscopic examinations at the 102 cm, 100 cm, 98 cm, 96 cm, 94 cm, or 92 cm resonant lengths tested. Under conditions of zero arterial flow, platelet and fibrin coagulum was detected adherent to the endothelium of the ligated artery or detached in the arterial lumen in 3 of 6 samples (Table 1 of accompanying manuscript). Under zero arterial flow conditions, no gross or histologic damage to the vascular walls and no delayed contrast enhancement on MRI were identified.

DISCUSSION

This study shows that no clinically significant RF-induced heating occurs during real-time SSFP MR imaging of MR-assisted catheter tip steering for interventional MRI. The maximum increase in temperature observed after 15 minutes of continuous SSFP imaging was 0.35 °C at 15 cm from magnet isocenter, which is well-below the 4 °C increase that can cause irreversible tissue damage[16,17]. As expected, temperature increases around the coils were directly proportional to the proximity of the catheter-coil tip to the magnet bore wall, as well as to the power of the MR pulse sequence (varied here by changing the RF flip angle).

Temperature rises caused by RF heating effects on the laser lithographed microcoil-catheter construct in air or saline also did not reach a significant level, with maximum tempurature rise reaching 1.84°C in the absence of saline coolant flow and with the microcoil-catheter tip positioned as far offset from isocenter as possible.

Following SSFP imaging after *in vivo* catheter placement, delayed contrast enhanced MRI revealed no evidence of vascular injury under normal flow and zero flow conditions. Histopathological evaluation of these carotid arteries confirmed the absence of vascular injury.

These findings indicate that RF heating previously observed in other types of conductive wires in MRI also occurs with microcoil-tipped endovascular catheters, however, to a lesser degree. The mechanism by which this RF induced heating occurs is likely due to the antenna effect, whereby the wire couples with the time-varying electric fields produced by the RF pulses. The amount of RF heating caused by conductive wires has been shown to be high in previous studies, although results have been variable and inconsistent. *In vitro* temperature increases in the range of 18 to 48°C have been reported, in experiments involving a standard conducting guidewire, coaxial cable, and miniature tracking coil[6,18], levels that are incompatible with patient safety[5-7]. These studies, however, are concerned with single wires whose distal end is an open circuit. In the case of microcoils used for catheter tip deflection, current carrying wires transit the length of the catheter lumen in pairs. Each pair is connected at the distal end to a conducting microcoil with low inductance compared to the coupled lead wires. This configuration is essentially a long, narrow loop antenna. Such an antenna is inefficient because the area enclosed by the loop is small. Indeed, the induced currents were found to be small so that significant cable resonances were not produced, or were damped out significantly at the wire length and field strength tested. There is no certainty that this solution will work at higher frequencies (such as 3T), or with different cable lengths. The results shown in Figure 5, with differential heating in air versus saline, suggest that resonances may play a small role under the conditions tested, although the larger heat capacity of saline as opposed to air may also explain these differences. In addition, a heating study involving intravascular imaging coils, which consisted of a loop with a length of 40 mm and a width of 6 mm, demonstrated much lower heating[19], probably due to the rounded end of the coil, which prevents a high concentration of electric fields. The coils used in MRI-assisted catheter steering are similar in size and shape to intravascular imaging coils.

Despite these promising results, further steps can and should be taken to make MRI-assisted catheter tip steering safer for clinical applications. Resistive heating could be reduced by a number of modifications to the catheter-coil construct. Increasing the efficiency of the catheter, i.e., increasing the amount of deflection obtained per unit of current, may involve the use of highly flexible materials (lower elastic modulus) for catheter construction. Since power dissipation is also proportional to the resistance, conducting wires of smaller resistance per unit length, such as thicker diameter wires, or gold wires, could also be used to reduce ohmic heating. Catheter bending stiffness can also be decreased by reducing the diameter and wall thickness of the catheter, thus reducing the area moment of inertia. More coil turns increase the magnetic moment but the wire width must also be reduced to accommodate more turns in the same overall space. Both the increase in turns/wire length and the reduction in wire width increase overall resistance and heating. Also, since the magnetic torque produced by this mechanism is the cross-product of the magnetic moment and the main scanner magnetic field, doubling the magnet strength to 3T was shown to double the effective catheter deflection[2]. This advantage, however, needs to be weighed carefully against the higher specific absorption rates (SAR) that are associated with higher field scanners since higher field strength may result in relatively more RF heating. The prototypical catheter-coil design used here involves winding of a magnet wire coil around the outside of the catheter. Catheter extrusion with the coils built into the walls of the catheter would increase insulation from the overlying blood and vessel walls.

The applied direct currents necessary to achieve catheter tip deflections result in resistive dissipation and more significant tissue heating than RF heating[20]. As discussed in the accompanying manuscript, infusion of normal saline at clinically achievable flow rates within the lumen of the catheter during application of current is successful in transferring heat produced by applied currents both *in vitro* and *in vivo*. Furthermore, the transfer of unwanted heat to coolant flowing in the catheter lumen is facilitated by using a high heat conductivity material as the catheter tip substrate on the inner surface of the coils, while using an insulating material on the outside of the coils to protect the blood and vessel walls. Using a clinically achievable saline coolant flow rate (10 ml/kg/hr), preliminary *in vivo* testing demonstrated no thermal injury to vessel walls from resistive or RF heating at physiologic and zero arterial flow. Although tissue temperature was not measured directly, the most important parameter in ultimate clinical application is the likelihood of thermal injury to tissues by the catheter either from resistive heating while current is applied for catheter navigation or RF heating while the catheter is being imaged in the MR scanner. At the resonant lengths between 102 cm and 92 cm from the femoral artery access site, there was no evidence for the massive RF-induced heating that has been reported in other guidewire constructs [4,6,11-13].

Although we have used saline coolant flowing through the catheter lumen to decrease temperature increases related to resistive heating, this approach could also be applied in the settings where RF heating is more significant. Similarly, the flowing saline coolant approach may provide device cooling at different cable lengths and at different MR field strengths than those tested, thus potentially providing a simple solution to heat dissipation under a variety of interventional MRI conditions. Several authors have investigated the use of coaxial chokes[3], transmission lines[21], and other devices along the catheter shaft, to reduce RF heating. The use of saline flowing through the lumen can be easily applied to almost any catheter system with a simple design modification. This approach would also be compatible with a wide variety of cable or catheter lengths and field strengths, without requiring complex design adaptations for each new device.

Although this is a promising result, further studies are necessary to examine the effects of varying resonant lengths of wire [22] to better characterize RF safety.

CONCLUSION

MRI-assisted steering coils for endovascular catheter steering during vascular intervention do not appear to experience significant RF heating. These results, combined with demonstration that resistive heating due to applied direct current can be minimized by transferring heat to saline coolant flowing through catheter lumen, begin to address safety concerns related to eventual clinical use of this device in vascular interventions and will serve as the basis for further *in vivo* testing under different scenarios of catheter navigation.

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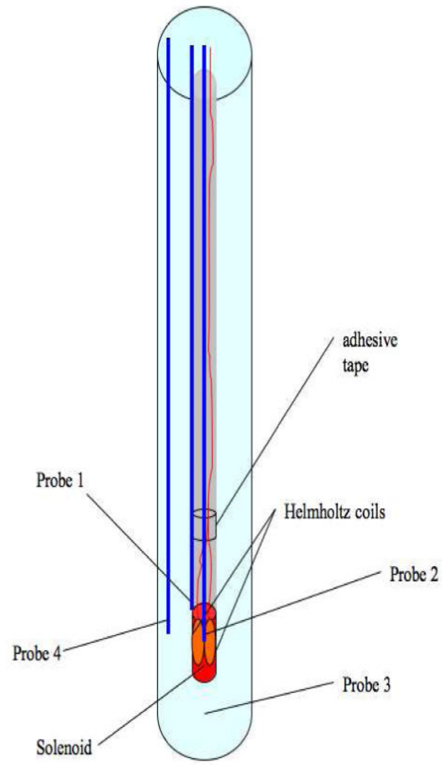

Figure 1.
Schematic of hand made coil-tipped catheter, gel phantom, and luxtron temperature probe set-up for RF heating experiments. The gel was allowed to solidify with the catheter in the center of the plastic tube phantom and the Luxtron thermometric probe tips were fixed in the positions indicated.

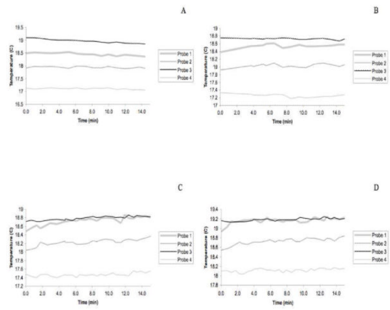

Figure 2.

Effect of distance from magnet isocenter on catheter tip temperature during real-time MR imaging. **A.** At isocenter, no significant heating was observed in any o the probes. **B.** 5 cm from isocenter, a slight rise in temperature was seen close to the catheter tip in probes 1 (0.20 °C) and 2 (0.14 °C). No significant increase in temperature was detected away from the catheter tip in probes 3 and 4. **C.** 10 cm from isocenter, the rise in temperature was 0.35 °C in probe 1 and 0.33 °C in probe 2. Again, no significant increase in temperature was detected away from the catheter tip in probes 3 and 4. **D.** 15 cm from isocenter, the increase in temperature was 0.30 °C in probe 1 and 0.30 °C in probe 2. Again, no significant increase in temperature was detected away from the catheter tip in probes 3 and 4.

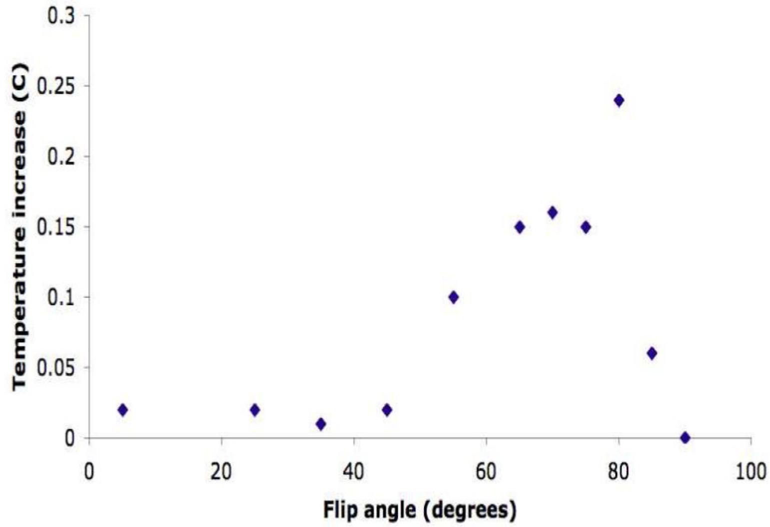

Figure 3.

Effect of flip angle. RF heating was observed at higher flip angles. After 10 min of continuous MR imaging, the heating was minimal – the highest observed increase in temperature over baseline was 0.25 °C with an 80° flip angle, measured in Probe 1.

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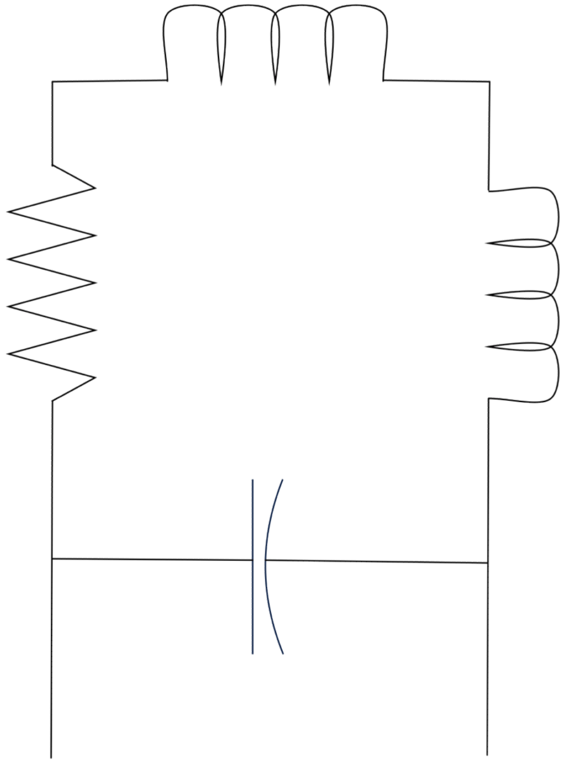

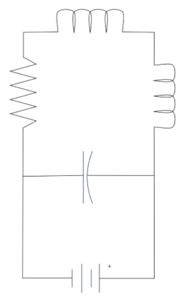

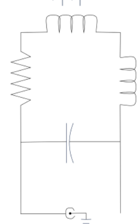

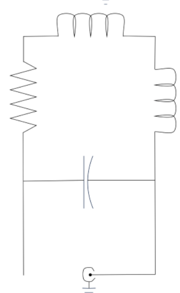

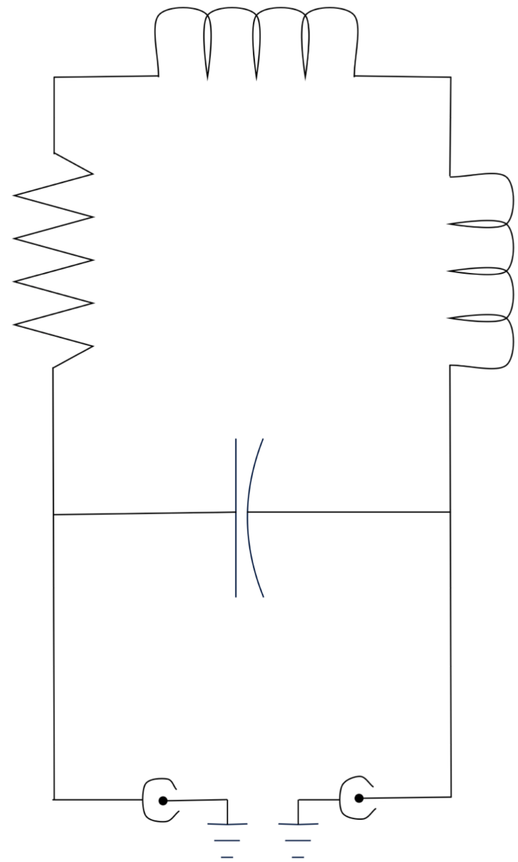

Figure 4.

Schematic of laser lithographed coil-tipped catheter and MR scanner for RF heating experiments. (A) Catheter leads not connected to any external ground (“floating”). (B) Catheter leads attached to power supply through a low pass filter, with no current applied. (C) Catheter leads short-circuited by grounding the outer layer of the power supply’s Bayonet Neill-Concelman (BNC) connector (C1), the inner wire of the BNC connector (C2), or both (C3).

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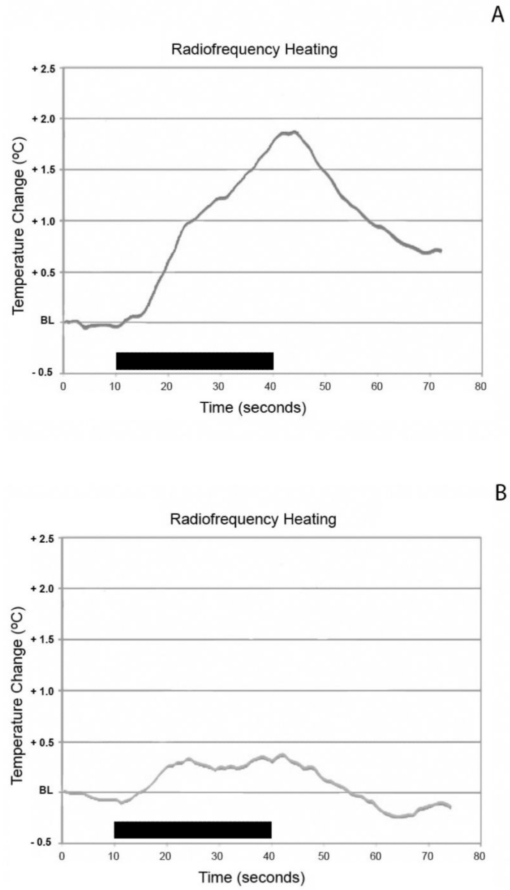

Figure 5.

Temperature rise measured by Luxtron fiber optic probe mounted on single axis Helmholz coil-tipped catheter placed in groove of 1.5 T magnet dovetail (maximally offset from center of bore, by 27.4 cm) under various grounding, filtration, and imaging schemes, as detailed in Table 1. Maximal temperature rise was in a single lead grounded construct in air (A). Minimal temperature rise was in an ungrounded construct suspended in saline (B). The period of RF exposure from second 10 to 40 is indicated by black rectangles at the bottom of each graph. Note that temperature increases during the 30 seconds the imaging sequence is active (seconds 10 to 40) and then declines rapidly after imaging ceases.

Table 1

Temperature rise measured by Luxtron fiber optic probe mounted on single axis coil-tipped catheter placed in groove of 1.5 T magnet dovetail (maximally offset from center of bore, by 27.4 cm) under various grounding and filtration schemes. Acquisition time consists of 10 seconds without imaging, followed by 30 seconds of SSFP imaging, followed by 30 seconds without imaging

SAR (Watts/kg)	Acquisition Time (sec)	Environment	Grounding or Filtration	Temperature Rise (°C)
4	30	Air	**Catheter wires not connected ("floating"** Fig 4A **)**	1.86
4	30	Air	**Short circuit; power supply on at zero current** (Fig 4B)	0.85
4	30	Air	**Short circuit; power supply turned off** (Fig 4B)	0.89
4	30	Air	**Inner wire grounded and outer layer BNC floating** (Fig 4C1)	1.53
4	30	Air	**Inner wire floating and outer layer BNC grounded** (Fig 4C2)	0.73
4	30	Air	**Inner wire and outer layer BNC grounded** (Figure 4C3)	1.91
4	30	Saline	**Inner wire grounded and outer layer BNC floating** (Fig 4C1)	0.45
4	30	Saline	**Inner wire and outer layer BNC grounded** (Figure 4C3)	0.55