RF Safety of Wires in Interventional MRI: Using a Safety Index

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With the rapid growth of interventional MRI, radiofrequency (RF) heating at the tips of guidewires, catheters, and other wire-shaped devices has become an important safety issue. Previous studies have identified some of the variables that affect the relative magnitude of this heating but none could predict the absolute amount of heating to formulate safety margins. This study presents the first theoretical model of wire tip heating that can accurately predict its absolute value, assuming a straight wire, a homogeneous RF coil, and a wire that does not extend out of the tissue. The local specific absorption rate (SAR) amplification from induced currents on insulated and bare wires was calculated using the method of moments. This SAR gain was combined with a semianalytic solution to the bioheat transfer equation to generate a safety index. The safety index (°C/W/kg) is a measure of the in vivo temperature change that can occur with the wire in place, normalized to the SAR of the pulse sequence. This index can be used to set limits on the spatial peak SAR of pulse sequences that are used with the interventional wire. For the case of a straight resonant wire in a tissue with very low perfusion, only about 100 mW/kg/°C spatial peak SAR may be used at 1.5 T. But for ≤10-cm wires with an insulation thickness ≥30% of the wire radius that are placed in well-perfused tissues, normal operating conditions of 4 W/kg spatial peak SAR are possible at 1.5 T. Further model development to include the influence of inhomogeneous RF, curved wires, and wires that extend out of the sample are required to generate safety indices that are applicable to common clinical situations. We propose a simple way to ensure safety when using an interventional wire: set a limit on the SAR of allowable pulse sequences that is a factor of a safety index below the tolerable temperature increase. Magn Reson Med 47:187–193, 2002. © 2002 Wiley-Liss, Inc.

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The goal of this work is to present such a model. This work will demonstrate a method by which insulated and uninsulated wire structures can be numerically analyzed to calculate the maximum amount of heating that can occur with the device. The safety index is introduced as a measure of RF safety that is independent of the specific geometric configuration or type of body coil used. The theory is based on wires that are completely inserted in the patient and is therefore directly applicable only to devices such as wire leads for pacemakers, neurostimulators, and spinal fusion stimulators. Still, many of the principles developed may apply to any wire structure used in MRI, including guidewires and catheter RF transceivers. The calculations in this study were performed for 1.5 T (64 MHz) and must be reevaluated for other field strengths.

THEORY

RF Heating in General

The determination of RF heating can be described as depicted in Fig. 1a. The input, P, is the time-averaged power of the applied RF pulse of the imaging sequence. This is converted into a distribution of deposited power, typically characterized by the specific absorption rate (SAR), which is a function of position, \( T \), in the sample. This SAR distribution is then transformed into a temperature distribution, \( \Delta T \), based on the thermal properties of the tissue.

Modeling

The first system is accurately modeled with Maxwell’s equations to find the electric field in the sample, which in turn gives rise to the SAR. The SAR distribution depends on the coil geometry and the electromagnetic properties of the tissue, including the electrical conductivity, \( \sigma \), and permittivity, \( \varepsilon \). SAR, given in units of W/kg, can be calcu-
lated from the electric field according to the following equation (9):

$$SAR = \frac{\sigma E^2}{2\rho_t}$$  \[1\]

where $E$ is the amplitude of the electric field, $\sigma$ is the electrical conductivity, and $\rho_t$ is the mass density of the tissue.

The second system is modeled with the bioheat transfer equation. We have previously shown that for the case of local RF heating in deep tissues, the bioheat equation can be solved by convolving the SAR distribution with the Green’s function of the bioheat equation (10). This approach, which can be thought of as a weighted spatial averaging, gives a physiological model-based prediction of thermal heating (10) that is more relevant than a simple arithmetic average over an arbitrary fixed volume as prescribed by current regulatory guidelines (11,12).

RF Heating With Conductive Wires

When a conductive wire structure is placed in the imaging sample, the electric field distribution is altered near the wire, resulting in an altered SAR distribution. We quantify this effect with an SAR gain, which is a function of position. Buechler et al. (13) used a similar measure (an electric field multiplier) to report calculations of induced electric fields near metal implants due to switched gradient magnetic fields. This adds a third system into the calculation procedure, as shown in Fig. 1b.

However, raw SAR gain alone is not a meaningful measure of the increased heating associated with an implanted wire, since bioheat transfer effects have not been accounted for. Therefore, the raw SAR gain was combined with the bioheat transfer convolution to produce a safety index. This safety index, which has units of °C/(W/kg), estimates the in vivo temperature increase that can occur with the wire in place for each unit of applied SAR. Therefore, it is helpful for setting limits on the peak SAR of pulse sequences that are used with the wire.

**METHODS**

**Calculation of SAR Gain**

The difficulty of finding SAR distributions surrounding an implanted wire is in calculating the induced current distribution on the implanted wire. Even for simple geometries, the problem cannot be solved analytically. Therefore, a numerical technique based on the method of moments (14) was used to find the induced currents on the wire. Atlamazoglou and Uzunoglu (15) have given a detailed description of the application of this procedure to finding induced currents and their resulting electric field distributions for both bare and insulated center-driven antennas in a homogeneous, infinite, lossy medium. In their case, the driving function was non-zero only at the driving point of the antenna.

We extend their work to a passive wire situated in an incident driving electric field by making one simple change: the driving function is changed from an impulse at the center of the wire to a driving function that exists all along the wire. A uniform parallel electric field that would generate a unit applied SAR was assumed as the driving field. The raw SAR gain, then, was equal to the magnitude of the resultant total SAR from the combined incident and induced electric currents.

**Experimental Verification of SAR Gain**

A phantom was constructed which enabled accurate and reproducible placement of the wire and temperature measuring probes in a gel to measure SAR. A schematic is given in Fig. 2. A cylindrical phantom (20 cm diameter, 60 cm length) was poured in two halves to allow access to a longitudinal plane. The phantom consisted of a polyacrylamide gel (6.5% acrylamide, 0.3% bisacrylamide, 0.05% TEMED, 0.08% ammonium persulfate; EM Science, Gibbstown, NJ) doped with 0.35% sodium chloride. To measure the complex permittivity of the gel, a small cylindrical sample of the gel was extracted and placed...
between two copper disks to form a parallel plate capacitor. The complex permittivity of the gel was calculated from the geometry of the capacitor and a measurement of the impedance of the capacitor using a network analyzer (Model 4195A; Hewlett Packard, Palo Alto, CA). Its relative permittivity was 77 and its conductivity was 0.5 S/m. These are representative values for human tissue between 10 and 100 MHz (16). It is necessary to use a gel to prevent convective heat transfer, which limits the achievable temperature gradients in a solution.

Silver-plated copper hookup wire (Alpha Wire, Elizabeth, NJ) was straightened and cut into pieces ranging from 3–39 cm in 3-cm increments. Three cases were tested: 0.8-mm-diameter (20 AWG) bare wire; 0.5-mm-diameter (24 AWG) bare wire; and 0.5-mm-diameter (24 AWG) wire with 25-μm (0.001-inch) thick insulation (polyester heat shrink tubing; Advanced Polymers, Salem, NH). Each wire was centered longitudinally in the phantom. A removable spacer was used to ensure that each wire was positioned at a uniform radial depth of 4.5 cm. This was deep enough to ensure that the phantom looked almost like an infinite medium from the wire surface but was shallow enough that there would be sufficiently high absolute heating to permit accurate SAR measurements. The phantom was positioned in the scanner such that the cylindrical gel was coaxial with the scanner bore.

Three custom-built fiber optic temperature probes (Fiso Technologies, Ste. Foy, Quebec, Canada) were connected to a data acquisition system to measure temperature changes at the center and both tips of the wire. This system had a measurement accuracy of 0.05°C and was capable of switching between channels every 0.2 s. This gave a temporal resolution of 0.6 s per probe. Each temperature probe provided an average temperature measurement over its cylindrical sensitive region that was 8 mm long and 0.6 mm in diameter. The sensitive region was configured in the center of a 30-cm polyimide tube (HV Technologies, Trenton, GA) to allow accurate positioning of the sensitive region at the tip of the wire. The thin wall of the tube (95 μm) did not appreciably affect the response of the temperature measurement. SAR was calculated by finding the initial slope of the temperature rise $\frac{dT}{dt}$ and multiplying by the specific heat capacity of the gel.

Experiments were performed on a GE Signa LX cardiac scanner. A pulse sequence optimized for high SAR deposition, rather than image quality, was used. Pulse sequence parameters were: fast gradient echo, 170° flip angle, and 11.4-ms effective TR. The estimated peak SAR given by the scanner was 4.0 W/kg. The transmit gain was increased 3 dB over the self-calibrated prescan level to double the applied power. Maximum SAR gain was computed as the ratio of the measured SAR at the tips of the wire to the SAR at the same spatial location in the gel without the wire present.

To directly compare theoretical and measured SAR gain, the theoretical SAR was spatially averaged with a thin disk to compensate for the spatial averaging caused by the temperature probes. The diameter of the averaging disk was 8 mm, equal to the length of the temperature probe’s sensitive region.

FIG. 3. Theoretically predicted SAR gain on the surface of a bare wire in a lossy medium subject to an incident homogeneous electric field. SAR gain is the ratio of peak SAR to the native SAR at the same location if the wire were not present (without Green’s function averaging). RF heating is minimum at the center of the wire and maximum at the wire tip in all cases. Five wire lengths are shown: 6 cm, 12 cm, 18 cm (close to resonant length), 24 cm, and 30 cm. Wire diameter is 0.5 mm (24 AWG) and electrical conductivity is 0.5 S/m.

Calculation of the Safety Index

Once the SAR gain model was experimentally verified in the phantom experiment, the safety index was calculated to produce a meaningful measure of the intrinsic safety of a given wire. Similar to Ref. 10, raw SAR gain distributions were convolved using MATLAB (The Mathworks, Inc., Natick, MA) with the Green’s function of the bioheat equation for a variety of wire types.

The safety index is the spatial maximum of the Green’s function convolved SAR gain distribution (°C/(W/kg)). It was verified that this spatial maximum always occurred on the axis of the wire at the wire tip. It was therefore only necessary to compute the convolution at this single point. This could be done with a 2D calculation. SAR gain and the Green’s function were sampled with increasing resolution and extent until doubling either parameter had a <1% effect on the final result.

The safety index was not experimentally verified because it incorporates in vivo bioheat transfer effects that are very difficult to replicate and control.

RESULTS

Wire Tip Heating—SAR Gain

Figure 3 shows theoretically predicted SAR gain distributions along the surface of a bare wire implant. Five examples are shown to demonstrate that wire heating is maximum at the wire tip for wire lengths shorter than, equal to, and greater than the resonance length.

Experimental Verification

Figure 4 shows an overlay of the theoretically predicted and measured SAR gain, accounting for the spatial averaging of the temperature probe, for each of the three tested
cases: 0.8-mm-diameter (20 AWG) bare; 0.5-mm (24 AWG) bare; and 0.5-mm (24 AWG) insulated wires. Changes in wire diameter slightly affect the magnitude of the SAR gain but do not shift the resonance peak. Insulating a wire reduces the magnitude of the resonance peak and causes it to shift to a longer length. Insulation also increases the length of short wire that exhibits virtually no SAR gain.

Safety Index

Figure 5 shows the dependence of the calculated safety index on the wire diameter, the electrical permittivity of the insulation, the insulation thickness, and the electrical conductivity of the tissue. This figure demonstrates how the resonance conditions (length of wire and amplitude of peak heating) are affected by each parameter. Changing the wire diameter does not shift the resonance peak, but alters the amplitude of the resonance peak. The permittivity of the insulation over the range of standard plastics ($\varepsilon_r = 2-4$) shifts the resonance length but does not change its amplitude. Increasing the insulation thickness causes the resonance length to get longer and the amplitude of that resonance to decrease. Increasing the electrical conductivity of the ambient medium flattens and broadens the resonance peak while shifting the resonance to shorter lengths. In each case, the same thermal parameters for the Green’s function were used ( thermal conductivity = 0.4 W/m°C, resting muscle perfusion = 2.7 ml/100 g/min (17)).

Figure 6 shows the dependence of the safety index on tissue perfusion for two cases: 1) a resonant bare wire, and 2) a short insulated wire. In each case, the safety index for the wire is shown in comparison to the wire-free safety index. The wire-free safety index is the SAR-to-temperature scaling factor for the case of uniform local heating (10). Since the safety index is a relationship between in vivo temperature and applied SAR, a safety index exists for the wire-free case that is a function of perfusion. In all cases, increasing the perfusion decreases the value of the safety index. A lower valued safety index indicates a safer condition.

DISCUSSION

Safe Length

The concentration of RF heating at the wire tips predicted by the theoretical model, shown in Fig. 3, is in good qualitative agreement with previous observations of wire tip heating (2–5).

As seen in Fig. 4, the resonance peak occurs at a length substantially less than half a wavelength and is broad. In this case, the resonance occurs between 15 and 18 cm, while the theoretical half wavelength is 21.5 cm. It has been assumed that wire lengths less than a quarter wavelength are generally safe (19,20). However, for the bare-wire case examined, significant SAR gain still exists at both the theoretical quarter wavelength and half the experimental resonant length. Figure 5b suggests that

$$\lambda = \frac{2\pi}{\sqrt{\mu / \varepsilon}} \sqrt{1 + \left(1 + \frac{\omega_0}{\omega_c}ight)^2}$$

Where $\lambda$ is wavelength, $\omega$ is angular frequency, $\mu$ is magnetic permeability of the medium, $\varepsilon$ is permittivity of the medium, and $\sigma$ is the electrical conductivity of the medium. Wavelength in the experimental phantom is 43 cm.
wires ≤ 10 cm will have a markedly reduced SAR amplification only when they are insulated with an insulation thickness that exceeds 30% of the radius of the wire.

Experimental Error

The theoretical prediction and experimental measurements in Fig. 4 show excellent agreement for the bare-wire cases. In the insulated case, the measured SAR gain is sometimes significantly lower than predicted. This is likely due to inconsistency in the insulation thickness at the tip of the wire. Insulation thickness along the length of the wire is easily controlled. However, wire tip heating is especially sensitive to the geometric configuration of the wire tip. Lower measured heating may have been caused by an excess of insulation at the wire tip. This suggests a potential strategy for designing safer wires by placing thicker insulation at the wire tips. The electric fields at the tip of the wire will thus reside in the lossless dielectric, reducing the power deposition in the lossy medium.

Exposed Wire Tips

During the insulated wire experiments, measured heating was sometimes up to 10 times greater than expected. This occurred when the tip of the wire was not completely sealed with the insulator. Since the predicted induced currents approach zero at the tips of the wire in the present model, this situation is not accounted for by the method of moments calculation. Chou et al. (8) similarly observed a 5.5-fold increase in RF heating at the tip of a broken spinal fusion stimulator lead compared to an intact one. We therefore stress that the method-of-moments model should not be applied to wires with exposed tips. Conversely, it is possible that insulated wire structures with exposed tips could be made safer if their tips were sealed.

Safety Index

The safety index takes into account both the SAR gain caused by introducing the wire as well as local SAR aver-
The advantage of using the safety index is that this provides an intrinsic measure of the relative safety of the interventional wire that is independent of the transmit coil or pulse sequence used. In other words, the safety index is a property of the wire and does not describe the total heating problem. The remaining part of the problem involves the pulse sequence and the transmit coil. Using the safety index to divide the problem into two parts allows the interventional wire and the transmit coil to be designed and optimized independently. The highest absolute RF heating that can be caused by the interventional wire is the product of the spatial peak SAR generated by the pulse sequence and the safety index of the interventional wire being used. Therefore, a simple way to ensure safety when using an interventional wire will be to place a limit on the SAR of allowable pulse sequences that is a factor of a safety index below the tolerable temperature increase.

For example, under worst-case conditions with a resonant wire in a tissue with very low perfusion (bone perfusion = 1.4 ml/100 g/min), the safety index can be as high as about 10°C/(W/kg). This means that if 1°C is the temperature limit, a pulse sequence with only 100 mW/kg spatial peak SAR may be used. For ≤10-cm wires that are placed in well-perfused tissues, and whose insulation thickness is at least 30% of the radius, the safety index can fall below 0.25°C/(W/kg). In this case, the whole-body SAR limits dominate and normal operating conditions of 4 W/kg spatial peak SAR are permissible. These quantitative results are only valid at 1.5 T (64 MHz). At other field strengths, and therefore frequencies, wavelength and tissue conductivity will be different. The method can be repeated for other field strengths with a change in these parameters.

Finally, since the safety will be determined by multiplying the scanner-estimated peak SAR by the safety index, errors in the peak SAR estimate will be magnified. Therefore, MRI system manufacturers must be conscientious in minimizing the error of these estimates.

Limitations

The present work has dealt exclusively with straight wires that are completely embedded in the sample subject to a uniform incident electric field. Therefore, the quantitative results must not be directly applied to curved wires or partially inserted wires (such as catheters and guidewires), or used with surface transmit coils that have highly non-uniform RF fields.

Despite these limitations, many of the main findings of this work may at least be qualitatively applicable to other cases. Future work can apply the method-of-moments theory to deal with curved wires. However, the analysis of partially inserted wires requires several extra degrees of complexity and may be better examined by another method.

CONCLUSIONS

We have demonstrated the first theoretical model of wire tip heating that can actually predict the absolute value of that RF heating and can therefore be used to formulate safety margins. We have introduced the safety index as a measure of the safety of an interventional wire that is independent of the particular MRI system with which it
will be used. This safety index acts as a convenient relationship between the spatial peak SAR of pulse sequences that will be used with the wire and the maximum tolerable in vivo temperature change.

Insulating wires that are shorter than half a wavelength (in the tissue) will improve their safety. The rule of thumb that wires less than a quarter wavelength (in the tissue) are safe is only true for wires whose insulation thickness is at least 30% of the radius of the wire. Finally, the quantitative RF safety indices given here are valid only at 1.5 T (64 MHz) and must be reevaluated for other field strengths.

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