Abstract—Magnetic Resonance Imaging (MRI) scans are contraindicated for many patients with medical implants. We establish the circumstances that cause, and the resistances required to ameliorate and to eliminate dangerous levels of MRI-induced heating that occur at the exposed, distal end of an electrical lead implanted in tissue. Simulated predictions are compared with measurements made at 128 MHz in a 3-Tesla MRI machine. A low resistance at kilohertz frequencies is sought by implant makers, in contrast with the high resistance demanded for safety. The practicality of presently-developed strategies to prevent tissue damage is brought into question. We examine the extent to which skin-depth and transmission-line properties can be manipulated to improve safety.

Index Terms—Biomedical electrodes, medical diagnostic imaging, dipole antennas, electromagnetic modeling.

I. INTRODUCTION

The strong magnetic field present in a Magnetic Resonance Imaging (MRI) machine presents a hazard, as most people know. Nevertheless, it is the energy delivered by the high-power RF field that is of greater concern for the patient. A 3-Tesla (3 T) MRI machine can deliver peak pulses in the order of 30 kW at 128 MHz. [1] This can lead to significant heating of the patient. The International Commission on Non-Ionizing Radiation Protection (ICNIRP) recommends maximum localized temperatures of 38 °C in the head and 39 °C in the torso, representing 1–2 °C of heating. [2] The dielectric heating of a patient is monitored by the Specific Absorption Rate (SAR), the power absorption per unit mass. The SAR is typically averaged over the whole-body, whole-head, or 10 g of mass.

The leads associated with medical implants can behave as antennas and concentrate the RF field. [3] We will show that this phenomenon is worst when the lead is just short of one-half wavelength long, about 250 mm for a typical lead in standard saline in a 3 T machine. There are claims in the literature that 10 g is too coarse near implants [4], and we confirm and exemplify this. While pacemakers have short leads, Spinal-Cord Stimulation (SCS) and Deep-Brain Stimulation (DBS) systems have leads that can exceed 600 mm in length because the implant is situated in the abdomen or chest cavity, far from the distal end of the electrode.

The MRI safety of implants has become a global issue and sites such as “Shellock” [5] are consulted throughout the world by MRI radiologists. Many implant-wareers are contraindicated for MRI scans. There is a growing need to address this problem. A spate of patents appeared in the last decade supposedly addressing the issue [6]–[11]. Only one product has appeared and it is rated for use only in 1.5 T machines, and only with a restricted scanning protocol. [12]

II. IMPACT OF A BARE WIRE

An implant lead wire acts as an “antenna in saline”. In a lossless medium (conductivity $\sigma = 0$), an ordinary dipole antenna will resonate when excited by an RF field when its length is approximately equal to an odd integer multiple of the half-wavelength: $0.5\lambda$, $1.5\lambda$, $2.5\lambda$, etc. In the conductive saline medium, the wave will compress and the wavelength becomes [13]:

$$\lambda = \frac{8\pi^2}{\omega^2 \mu \epsilon} \left[1 + \left(\frac{\sigma}{\omega \epsilon}\right)^2\right]^{-1/2}$$

(1)

where $\omega$ is the angular frequency, and $\mu$ and $\epsilon$ are the permeability and permittivity of the medium, respectively. Furthermore, the induced antenna currents are not confined by the ends of the wire and are free to extend into the medium. [14] The antenna will be effectively lengthened by the medium itself.

This situation leads us to expect heating to occur at the ends of a wire when stimulated at a frequency slightly below that at which the lead wire is one-half wavelength long. To confirm this expectation we simulate using COMSOL Multiphysics 4.4 with a phantom model set according to ASTM F2182-11a [15] with the exception of the length parameter, which is doubled from 0.6 m to 1.2 m, to accommodate the testing of longer length wires. For simulation details refer to [16].

Fig. 1 shows the simulated field at the end of a bare wire embedded in a human-body phantom and exposed to a calibrated SAR of 1 W/kg against length normalized to the wavelength via (1). Simulation confirms a strong resonance at $0.41\lambda_{Pn}$, confirming theoretical expectation.

III. IMPACT OF INSULATED WIRE WITH DISTAL END EXPOSED

Now we consider a wire with insulation covering all but one end, representing an implant’s distal electrode. The average electric field near the distal electrode at various wire lengths is in Fig. 2. The length is normalized to $\lambda_{Pn} = 0.61$ m and the first resonant peak occurs at $0.41\lambda_{Pn}$. The insulation shifted...
Fig. 1. Average E-field at one end of an 800 µm diameter bare wire for various lengths in a lossy phantom ($\sigma = 0.47 \text{ S/m}$) to model a human body. The wire length is normalized to a wavelength of 0.24 m.

Fig. 2. Average E-field at the distal electrode of an insulated wire in a lossy phantom ($\sigma = 0.47 \text{ S/m}$). The insulation covering the 800 µm diameter wire is 350 µm thick. The wire length is normalized to a wavelength of 0.61 m.

the resonant frequency and it has also increased the electric field intensity at the first resonant peak threefold.

The change in temperature ($T - 37 \degree C$) at the distal electrode of the insulated wire after 5 min of MRI scanning is considered next. The result with and without blood perfusion is shown in Fig. 3, again with the $x$-axis normalized to the wavelength of the electric field within the phantom, $\lambda_{Pn}$.

The inclusion of blood perfusion provides an approximation to living human tissue, reducing the heating by 18% at the worst-case length. At 0.41$\lambda_{Pn}$ (0.25 m), normal body temperature is exceeded by almost 22 °C, well over the 2 °C limit for SCS. The peak temperature declines as longer wire lengths but still mostly exceeds safety limits. A two-dimensional slice of the temperature $T$ in the $x$-$z$ plane at $y = 0$ is shown in Fig. 4, for a wire length of 0.41$\lambda_{Pn}$. This figure makes it clear that a sample size of 10 g is indeed too coarse for estimating SAR—10 g is a sphere of one inch diameter, large compared to the hot region of the plot.

IV. COMPARISON WITH MEASURED VALUES

In order to have confidence in our simulations, a torso-and-head phantom after [15] was built in clear acrylic and filled with 28 L of saline gel with a conductivity of 0.47 S/m. The ratio of NaCl and polyacrylic acid (PAA) to distilled water was 1.32 g/L and 10 g/L, respectively. Wire samples in the phantom were tested within a commercial 3T MRI scanner. Fig. 5 shows the phantom on the MRI machine bed.
Temperature was measured using an Optocon Fotemp GaAs-based fiber-optic system with TS2 probes. Fig. 6 depicts a temperature sensor tip aligned with the exposed tip of an 800 µm Cu wire of length 0.41λ, coated with 350 µm of plastic insulation.

The temperature rise after 5 minutes of MRI scanning was recorded for various lengths of insulated wire. The insulation was 350 µm thick, but thinner and thicker variants were also tested. The measurements are shown in Fig. 7 with simulated data overlaid. We measure a temperature rise of 7.9°C near the worst-case length, with an estimated whole-body SAR of 2.7 W/kg. We scale our measurements by 2.7 times to account for the conservative SAR estimate reported by the MRI machine. This corresponds to a rise of 21.3°C for a whole-body SAR of 1 W/kg. Similar wire with double the thickness of insulation produces almost 25.7°C, while just 21 µm of insulation produces only 3.7°C, consistent with our expectations in section VII.

V. IMPACT OF WIRE RESISTANCE

Implant manufacturers seek a low resistance connection to deliver therapeutic current and maximize battery life. An SCS implant typically delivers a maximum current of 10 mA. A resistance of 50 Ω/m is desirable, double that barely comfortable.

A reduction in the heating of the tissue near the distal electrode could be achieved with uniformly distributed resistance in the implant wire. Three-dimensional simulations of a hollowed out version of the implant wire with 350 µm of insulation were performed. The resistance was varied by controlling the resistivity of the metal. To ensure current uniformity, the resistivity was constrained to values such that the skin depth exceeded the thickness of the metal by at least 10 times. The resistance at 128 MHz and the temperature change after 5 min as a function of the metal resistivity is shown in Fig. 8. A 2°C limit would require >850 Ω/m of resistance. The significant benefit of a factor of two would require >200 Ω/m.

VI. IMPACT OF SKIN EFFECT

Is it possible to select values for the radius r and resistivity ρ of the wire such that the skin effect provides high enough resistance at 128 MHz and low enough at dc? Skin depth δ ≈ \sqrt{2\rho/\omega \mu} [17], about 14.5 µm for Pt. Resistance of a wire \( R = \rho/A \) Ωm, where A is effective cross-sectional area. For dc, \( A = \pi r^2 \), but smaller for increasing frequency, and approximated as \( 2\pi r \delta \) if \( r \gg \delta \). It is clear that dc resistance is inversely proportional to \( r^2 \) and to first order ac resistance to r. Thus larger \( r \) will give a larger ratio of ac to dc resistance, but falling total resistance. To have no more than \( R_{dc} = 50 \Omega/m \) in Pt requires \( r \gtrsim 26 \mu m \). Simple approximations for the
A coaxial transmission line is with conductive tissue representing the outer cylindrical conductor. The implanted lead wire resembles a coaxial transmission line but of the insulation leads to the results shown in Fig. 7 and 8. Skin depth may help, but it cannot deliver the required ratio.

VII. IMPACT OF INSULATION PROPERTIES

Varying the thickness, $t_i$, and the dielectric constant, $\epsilon_r$, of the insulation leads to the results shown in Fig. 7 and Fig. 9. These results can be understood by realizing that an implanted lead wire resembles a coaxial transmission line but with conductive tissue representing the outer cylindrical conductor. [19] The ideal characteristic impedance of a uniform coaxial transmission line is

$$Z_0 = \frac{1}{2\pi} \sqrt{\frac{\mu}{\epsilon}} \cdot \ln \left( \frac{r_2}{r_1} \right) \quad (2)$$

where $r_1$ and $r_2$ are the radii of the inner and outer conductors, and $\mu$ and $\epsilon$ are the absolute permeability and permittivity of the dielectric material, respectively. Its $Z_0$ affects the amount of RF energy that is coupled to the implant lead wire. It is clear that thin insulation, while potentially not practical for reasons of durability, offers a significant advantage. An insulation with a high dielectric constant will also help.

VIII. CONCLUSION

We explain the mechanism behind the hazard presented by implant electrodes, and confirm that 10 g is too coarse a sample over which to average exposure. Simulated and measured results agree. We have established lead resistance values required to reduce or eliminate the heating hazard at 128 MHz, and to preserve SCS implant battery life for low-frequency signals. We show that the skin effect alone is unable to provide the impedance difference. We show that manipulation of the insulation properties can lower peak distal heating by a large extent. In combination with other techniques that we are not free to disclose at this time [20], can be expected to lead to an MRI-safe electrode design.

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