Influence of Metallic Shielding on Radio Frequency Energy-Induced Heating of Leads With Straight and Helical Wires: A Numerical Case Study

Mikhail Kozlov, Member, IEEE, Wolfgang Kainz, Member, IEEE, and Luca Daniel, Member, IEEE

Abstract—Heating induced by radio frequency (RF) energy may appear in tissues near implants located in a human subject undergoing magnetic resonance imaging examination. Lead shielding was proposed to reduce such heating below dangerous levels. In this article, we employed 3-D electromagnetic (EM) and thermal co-simulations to quantify the effectiveness of metallic shielding in the presence of two large sets of nonuniform incident fields at 128 MHz. Specifically, we used a lead EM model (LEM) and computed the RF responses, i.e., the net dissipated power and net temperature increase, above background, at the electrodes. We considered a set of single electrode leads with both straight and helical wires, comparing regular unshielded configurations to different implementations of a metallic shield. The lead length, the relative permittivity of the insulator material, and the lead electrode geometry were all independently varied. For leads with helical wires we observed that: 1) the metallic shield significantly modified the LEM; 2) in most cases, the RF responses substantially increased if metallic shielding was applied; and 3) the net temperature increase in close proximity to the shield can be substantially higher than the net temperature increase in close proximity to the lead electrode. Leads with helical and straight wires exhibited significantly different behavior. Using the results obtained from generic straight wire leads to predict the results for leads with helical wires can be significantly misleading.

Index Terms—Active implantable medical devices (AIMDs), biomedical applications, electromagnetic (EM) modeling, EM shielding, radio frequency (RF) heating, RF safety, transfer functions (TFs).

I. INTRODUCTION

MAGNETIC resonance imaging (MRI) scanners use a strong static magnetic field, fast alternating magnetic field gradients, and a high level of radio frequency (RF) radiation to generate images of a human subject. RF energy absorbed by a human subject undergoing MRI examination could raise the local tissue temperature by more than 1 °C [1].

In human subjects with an active implantable medical device (AIMD), additional RF energy-induced heating is observed in the tissues near the lead electrodes [2]–[5], potentially causing tissue damage.

The presence of an AIMD results in: 1) scattering of the electromagnetic (EM) field by all surfaces of the AIMD and 2) intercepting of some RF power by the AIMD structures. A part of the intercepted RF power produces an electric current in conductive components of the AIMD. This current generates a back-radiated EM field projecting into the surrounding AIMD tissues. The scattered and back-radiated EM fields of the AIMD structures can be scattered backwards at the tissue interfaces. Thus, EM field multiple scattering can appear. The total EM field, i.e., the sum of the incident EM field, the back radiated EM field, and the multiple scattered EM field, differs substantially from the EM field in the absence of an AIMD. The sum of incident and multiple scattered EM fields alone usually does not result in an appreciable increase of local tissue temperature, compared to the levels observed without an AIMD. The primary concern is rather the heat generated by the back-radiated field that is deposited mostly in small volumes of tissue near the electrodes of the lead or near an edge of a conductive structure.

AIMD lead designs attempting to reduce the RF-induced heating during MRI examinations include two major categories: 1) modification of the implant structure to reduce the incident field on the lead wires and 2) mitigation of the amount of intercepted power that is back radiated from the AIMD electrodes. For a fixed geometrical AIMD electrode, the latter category includes a variety of approaches: modification of the electrical properties of the lead insulation [6], [7]; use of RF traps or chokes [8], [9], [10]; use of high-resistance or high-inductance leads [5], [11]; and altering the geometry of the lead wire [12]. The first category includes approaches such as adding a set of extra wires to be incorporated externally along the implant lead [13] and adding shielding [14]. Most of the approaches mentioned above can be combined.

The exact values of the static magnetic field and the RF are intentionally set to be different across all MRI scanner manufacturers. For example, the RF of the Siemens and Philips 3T scanners are 123.2 and 127.7 MHz, respectively. Thus, from an RF point of view, a solution to reduce the RF-induced heating during the MRI examinations should be wideband. Other important limiting factors of an AIMD design are: 1) the bio-compatibility of all AIMD materials that can be...
in contact with human tissues and 2) the long-term stability of the mechanical and electrical properties.

Shielding is widely used in a variety of non-MRI applications. Being relatively simple to implement, and inexpensive, it is considered as rather promising also for MRI. However, these considerations are based on studies that covered: 1) a small number of leadwire geometries and lengths and 2) a limited number of incident EM field distributions.

In our article [15] presented at the IEEE MTT-S International Conference on Numerical Electromagnetic and Multiphysics Modeling and Optimization, we developed a numerical workflow to investigate the influence of a metallic shield on the RF-induced heating of leads with straight and helical wires for a variety of incident RF electric fields, tangential to the AIMD lead. In that study, we assessed one lead length (320 mm), and one electrode geometry.

This article includes significant extensions to the workflow in [15], as well as further results: 1) an analysis of the RF-induced heating in close proximity to the metallic shield; 2) assessment of a wide range of lead lengths; 3) examination of different electrode geometries; and 4) examination of the metallic shield when fully in contact, length-wise, with the surrounding medium.

II. METHODS AND PROCEDURES

Quantities of interest were the RF responses: 1) the net dissipated power \(P\) around the lead electrode, and 2) the net temperature increase \(\Delta T\) at the electrode or shield

\[
P = \int_{HSIV} (\sigma \cdot |E_{\text{total}}(v)|^2 - \sigma \cdot |E_{\text{backgnd}}(v)|^2) \, dv
\]

\[
\Delta T = \Delta T_{\text{total}} - \Delta T_{\text{backgnd}}
\]

where \(\sigma\) is electrical conductivity of the surrounding medium, \(E_{\text{total}}(v)\) is the electrical field with the lead in place, \(E_{\text{backgnd}}(v)\) is the electrical field without the lead in place, \(HSIV\) is the hot spot integration volume, \(\Delta T_{\text{total}}\) is the temperature increase with the lead in place, and \(\Delta T_{\text{backgnd}}\) is the temperature increase without the lead in place.

A. Lead Description

The exact dimensions and geometric details of commercially available AIMD leads have not been published. We, therefore, relied on information from the literature, as well as publicly available pictures and datasheets to develop the electrode models used in this article. Our set of several generic leads included leads with straight and helical titanium alloy wires, with and without a titanium alloy shield. Two electrode geometries were investigated: 1) a “generic” electrode with a diameter of 0.25 mm and a length of 10 mm, and 2) a helical electrode made of 0.2 mm × 0.2 mm rectangular wire with a pitch of 0.9 mm, and an external diameter and length of 1.2 and 1.8 mm, respectively. The latter electrode dimensions were based on information from a publicly available specification sheet of the Medtronic Lead Model 3830 [16]. The diameter of the unshielded lead insulation was 1.46 mm. The straight wire diameter was 0.73 mm. The helical structure was a 0.1 mm × 0.1 mm rectangular wire with a pitch of 0.33 mm, and with an external diameter of 0.9 mm [see Fig. 1(a) and (b)]. The proximal lead end was capped with an insulator. The length of the leads was varied from 60 mm to 600 mm in 10-mm steps.

The outer diameter of the shield was 1.56 mm. A lead distal end section of 3 mm was not shielded. Four cases were studied and summarized in Table I: 1) lead without any shield (i.e., an unshielded lead); 2) entire external shield surface covered by an insulator with an outer diameter of 2.5 mm (i.e., an insulated shield); 3) an electrical contact between the insulated shield and surrounding medium through only a small section of 3 mm in length at the distal end (i.e., a mostly insulated shield) (see Fig. 2); and 4) the entire external shield surface in contact with the surrounding medium (i.e., a non-insulated shield). The impedance between the inner wire and the external metallic shield was defined as infinity at the proximal end.

Silicone elastomers are considered as more suitable for medical implants than carbon-based plastics and rubber, in many aspects. Due to the large variety of silicone elastomers, their
values of relative permittivity ($\varepsilon_r$) are in the range from 2.7 to 6 [17]. In our article, we selected two values within the range 2.7–5.5. The investigated matrix of alterable lead design parameters is presented in Table II. It resulted in eight possible case combinations of lead geometry designs. Taking into account the variety of shield designs, 32 lead design variants were investigated.

### B. Lead EM Model (LEM) Evaluation

In order to be able to evaluate the RF responses in humans to a set of clinically relevant $E_{\text{tan}}(l)$, the analytical LEM was adopted from [18]

$$P = A \cdot \left| \int_0^L S(l) \cdot E_{\text{tan}}(l) \, dl \right|^2$$

$$\Delta T = A_T \cdot \left| \int_0^L S(l) \cdot E_{\text{tan}}(l) \, dl \right|^2$$

where $A$ and $A_T$ are calibration factors, $S(l)$ is the transfer function (TF) at different positions, $l$, along the lead, and $L$ is its total length. The TF was the same for $P$ and $\Delta T$ evaluation [11] because the TF is a relative measure of different lead segment contribution to the EM field, radiated by a given electrode. The TF for each lead was numerical, obtained at 161 equidistant points applying the reciprocity approach detailed in [11]. The TF was normalized such that $\int_0^L |S(l)| \cdot dl = 1$. The phase at the electrode tip was shifted to $\phi(S(0)) = 0$. A calibration factor evaluation and a validation of the TF were performed using a linear regression analysis, based on the 3-D EM and thermal co-simulation results obtained for a set of artificial heterogeneous $E_{\text{tan}}(l)$ as detailed in [19].

Two sets of 40 heterogeneous $E_{\text{tan}}(l)$ were obtained using our numerical approach detailed in [11], [19], and [20]. The lead was positioned parallel to the $z$-axis in the middle of a box (600 mm × 400 mm × 2400 mm) filled with a medium whose electrical properties were $\varepsilon_r = 78$, and with an electrical conductivity of 1.2 S/m (similar to blood). Four numerical antennas were located along one ($yz$) side of the box, generating an EM field at 128 MHz so that the $z$-component of the electric field was dominant at the location of the lead. $E_{\text{tan}}(l)$, $E_{\text{backgnd}}(b)$, and $\Delta T_{\text{backgnd}}$ were modeled using the same setup, but without the lead in place.

### C. Modeling Convergence

The computational meshes of the 3-D EM and thermal numerical domains were independently generated in each solver to ensure the best suitable mesh for each simulation modality. $E_{\text{tan}}(l)$, $S$, $P$, and $\Delta T$ convergence was better than 3% and was ensured using the approach outlined in [19]. The 3-D EM mesh size in the region of maximum power deposition was less than 10 μm. The size of the mesh elements in the region of maximum thermal gradient was less than 8 μm. Comparison of numerical prediction and measurement results that provide a proof of validity of our 3-D EM, and thermal
co-simulation workflow was recently published in a separate article [21].

III. RESULTS AND DISCUSSION

A. 3-D EM and Thermal Co-Simulation Results

A rapid decrease of power loss density, i.e., \( \sigma \cdot |E_{\text{total}}(t)|^2 \), in close proximity to the electrodes was observed (see Fig. 4). Temperature profiles changed significantly in the first 10 s of transient heating (see Fig. 4) for the helical electrode. Significant heating was observed in areas with negligible power loss density levels (see Fig. 4). The straight wire operated as a thermal absorber. This resulted in relative cooling of the area close to the straight wire, especially for excitation times longer than 60 s. Because absolute heating of electrodes with helical wire leads was significantly smaller than for leads with straight wires, after excitation time of 900 s, the background heating contribution was substantially noticeable (see Fig. 5).

B. Comparison of the LEMs

The TFs for leads of 320 mm in length with straight electrodes were already presented in our article [15]. TFs for leads of length 80, 160, 320, and 480 mm, with helical electrodes, are new results shown in Figs. 6–9. The arrows in all figures point to the \( y \)-axis of the corresponding curve.

The lead electrode geometry visibly influenced the \( S(l) \) amplitude of the leads with helical wires. For leads with the straight wire, a change of the lead electrode geometry resulted in less than 5% variation on the \( S(l) \) amplitude. The shield significantly modified the TF magnitude and phases for the unshielded lead with a helical wire. For the same lead electrode and insulator \( \varepsilon_r \), the TF was very similar (difference less than 12%) for the following lead configurations: 1) leads with straight wires and shield designs: no shield, insulated shield, and mostly insulated shield and 2) for leads with helical wires and shield designs: insulated shield, and mostly insulated shields. For the same lead electrode and insulator \( \varepsilon_r \), the similar observations were derived: 1) for leads with straight wires and non-insulated shields and 2) for leads with helical wires and non-insulated shields.

The TF was successfully validated for all electrodes. Most \( R^2 \) values were close, in a range from 0.95 to 1. Only for
Fig. 7. TF amplitudes and phases for leads with helical wire, helical electrode, and insulator $\varepsilon_r = 5.5$, lead lengths of 80, 160, 320, and 480 mm.

Fig. 8. TF amplitudes and phases for leads with straight wire, helical electrode, and insulator $\varepsilon_r = 2.7$, lead lengths of 80, 160, 320, and 480 mm.

Fig. 9. TF amplitudes and phases for leads with straight wire, helical electrode, and insulator $\varepsilon_r = 5.5$, lead lengths of 80, 160, 320, and 480 mm.

Fig. 10. Calibration factor, $A$, for different lead design and length with helical electrode.

C. Influence of Shield Design on Net Tip Power Deposition

Direct evaluation of the influence of shield design on net power deposition was done using a ratio between the maximum $P$, observed for the shielded and the unshielded lead, when applying a given set of nonuniform $E_{tan}(l)$ (see Figs. 11–14). Two sets of 40 nonuniform $E_{tan}(l)$ were applied.
For leads with a helical wire, the major observations were similar for both $E_{\tan}(l)$ sets: 1) the insulated and mostly insulated shield, which was expected to reduce $P$, actually significantly increased the $P$, by more than one order of magnitude; 2) the ratio for the non-insulated shield leads substantially (up to 400%) depended on the electrode design and the insulator $\varepsilon_r$; and 3) for all types of investigated shields the ratio depended non-linearly on the lead length.

For the leads with straight wires the observations were as follows: 1) only the insulated shield slightly increased $P$ (up to 25%) for some lead lengths; 2) most often, the insulated shield decreased $P$ more than for 50%, with moderate dependence of the ratio on the lead length; and 3) the non-insulated shield decreased $P$ more than one order of magnitude for a lead length longer than 120 mm.

### D. Net Temperature Rise at Shield

To evaluate the $\Delta T$ level at the shield ($\Delta T_{\text{shield}}$) for shielded leads, relatively to $\Delta T$ level at electrode ($\Delta T_{\text{electrode}}$) of unshielded leads, we used the ratio between the maximum of $\Delta T_{\text{shield}}$ (max($\Delta T_{\text{shield}}$)) and the maximum of $\Delta T_{\text{electrode}}$ (max($\Delta T_{\text{electrode}}$)) over given $E_{\tan}(l)$ set (see Figs. 15 and 16).

For the insulated shield, the difference between $\Delta T_{\text{total}}$ and $\Delta T_{\text{backgd}}$ at the shield was less than 3% of $\Delta T_{\text{backgd}}$. Thus, leads with insulated shields were excluded from this analysis. The lead wire design had a significant influence on the ratio max($\Delta T_{\text{shield}}$)/max($\Delta T_{\text{electrode}}$). For leads with helical wires:
Fig. 16. Leads with straight wire and the first set of $E_{int}(t)$. Influence of shield design on the ratio $\Delta T_{\text{shield}}/\Delta T_{\text{electrode}}$ observed for different lead lengths, electrode designs, and insulator $\varepsilon_r$.

1) for mostly insulated leads the ratio was higher than 10; 2) for non-insulated shield leads the ratio was, in most cases, higher than 1; and 3) only for non-insulated shield leads with length shorter than 140 mm, the helical electrode, and insulator $\varepsilon_r$ of 2.7, the ratio was less than 1. For leads with straight wires: 1) the ratio was less than 1 for all leads with length longer than 110 mm and 2) the ratio was higher than 1 only for non-insulated shield leads with length shorter than 110 mm. Reduction of $P$ and $\Delta T$ at the electrode, through the use of a shield, makes no sense for patient safety if $\Delta T$, in close proximity to any part of the shield, is above the limit.

IV. CONCLUSION

Our numerical case study indicates that the LEM can be successfully applied to predict the net dissipated lead tip power $P$ around AIMD lead electrodes, as well as the net temperature increases at the electrode $\Delta T_e$ for leads with helical and straight wires and different types of shields. However, it is impossible to estimate the quantitative variation of $P$ and $\Delta T$ for different lead designs based only on the known maximum deviation of the TF, and the discrepancy of calibration factors. Utilization of the results obtained for generic leads with a straight wire can significantly mislead the prediction of results for leads with helical wires. For leads with helical wires, the net temperature increase, $\Delta T_e$, in close proximity to the shield, was sometimes observed to be higher than $\Delta T$ in close proximity to the lead electrode. An analysis of the shield influence on RF energy-induced heating of leads shall be performed for all possible lead lengths, electrode geometries, and insulator $\varepsilon_r$.

DISCLAIMER

The mention of commercial products, their sources, or their use in connection with material reported herein is not to be construed as either an actual or suggested endorsement of such products by the Department of Health and Human Services.

REFERENCES

[1] Z. Wang, J. C. Lin, W. Mao, W. Liu, M. B. Smith, and C. M. Collins, “SAR and temperature: Simulations and comparison to regulatory limits for MRI,” J. Magn. Reson. Imag., vol. 26, no. 2, pp. 437–441, Aug. 2007.

[2] J. B. Erhardt et al., “It’s the little things: On the complexity of planar electrode heating in MRI,” NeuroImage, vol. 195, pp. 272–284, Jul. 2019.

[3] T. Song, Z. Xu, M. I. Iacono, L. M. Angelone, and S. Rajan, “Retrospective analysis of RF heating measurements of passive medical implants,” Magn. Reson. Med., vol. 80, no. 6, pp. 2726–2730, Dec. 2018.

[4] S. Park, R. Kamondetdacha, A. Amjad, and J. Nyenhuis, “MRI safety: RF-induced heating near straight wires,” IEEE Trans. Magn., vol. 41, no. 10, pp. 4197–4199, Oct. 2005.

[5] P. Nordbeck et al., “Reducing RF-related heating of cardiac pacemaker leads in MRI: Implementation and experimental verification of practical design changes,” Magn. Reson. Med., vol. 68, no. 6, pp. 1963–1972, Dec. 2012.

[6] L. Golestanirad et al., “Reducing RF-induced heating near implanted leads through high-dielectric capacitive bleeding of current (CBLOC),” IEEE Trans. Microw. Theory Techn., vol. 67, no. 3, pp. 1265–1273, Mar. 2019.

[7] M. Kozlov and G. Schaefer, “Influence of electrical properties of lead insulation on radio frequency induced power deposition during magnetic resonance imaging at 64 MHz,” in Proc. Biomed. Eng., 2016, pp. 228–235, doi: 10.21366/2016.832-023.

[8] P. A. Bottomley, A. Kumar, W. A. Edelstein, J. M. Allen, and P. V. Karmarkar, “Designing passive MRI-safe implantable conducting leads with electrodes,” Med. Phys., vol. 37, no. 7, pp. 3828–3843, Jun. 2010.

[9] G. H. Griffin, K. J. T. Anderson, and G. A. Wright, “Miniaturizing floating traps to increase RF safety of magnetic-resonance-guided percutaneous procedures,” IEEE Trans. Biomed. Eng., vol. 64, no. 2, pp. 329–340, Feb. 2017.

[10] R. Das and H. Yoo, “RF heating study of a new medical implant lead for 1.5 T, 3 T, and 7 T MRI systems,” IEEE Trans. Electromagn. Comput., vol. 59, no. 2, pp. 360–366, Apr. 2017.

[11] M. Kozlov, M. Horner, and W. Kainz, “Modeling radiofrequency responses of realistic multi-electrode leads containing helical and straight wires,” Magn. Reson. Mater. Phys. Biol. Med., to be published, doi: 10.1007/s10334-019-00793-9.

[12] C. Jiang et al., “Deep brain stimulation lead design to reduce radio-frequency heating in MRI,” Electron. Lett., vol. 50, no. 25, pp. 1898–1900, Dec. 2014.

[13] S. Mecabe and J. Scott, “A novel implant electrode design safe in the field of MRI scanners,” IEEE Trans. Microw. Theory Techn., vol. 65, no. 9, pp. 3541–3547, Sep. 2017.

[14] K. Singhal and J. A. Nyenhuis, “Analysis and design of lead wires with metallic shielding for reduction of RF heating during MRI for active implants,” in Proc. 26th Annu. ISMRM Conf., 2018, p. 1454.

[15] M. Kozlov, W. Kainz, and L. Daniel, “Influence of a metallic shield/insulator $r_e$ on RF-induced heating near straight wires,” IEEE Trans. Magn., vol. 37, no. 7, pp. 3828–3843, Jun. 2010.

[16] L. Yu and A. L. Skov, “Silicone rubbers for dielectric elastomers with improved dielectric and mechanical properties as a result of substituting silica with titanium dioxide,” Int. J. Smart Nano Mater., vol. 6, no. 4, pp. 286–289, Oct. 2015.

[17] S.-M. Park, R. Kamondetdacha, and J. A. Nyenhuis, “Calculation of MRI-induced heating of an implanted medical lead wire with an electric field transfer function,” J. Magn. Reson. Imag., vol. 26, no. 5, pp. 1278–1285, Nov. 2007.

[18] M. Kozlov and W. Kainz, “Lead electromagnetic model to evaluate RF-induced heating of a coax lead: A numerical case study at 128 MHz,” IEEE J. Electron Magn., RF, Microw. Med. Biol., vol. 2, no. 4, pp. 286–293, Dec. 2018.

[19] M. Kozlov, M. Horner, and W. Kainz, “Modeling RF-induced power deposition and temperature rise of coaxial leads with helical wires,” in Proc. 41st Annu. Int. Conf. IEEE Eng. Med. Biol. Soc. (EMBC), Jul. 2019, pp. 1895–1898, doi: 10.1109/EMBC.2019.8856572.

[20] M. Kozlov, L. M. Angelone, and S. Rajan, “Effect of multiple scattering on heating induced by radio frequency energy,” IEEE Trans. Electromagn. Comput., to be published, doi: 10.1109/temc.2019.2950170.