Published in final edited form as:

Magn Reson Med. 2018 February ; 79(2): 1135–1144. doi:10.1002/mrm.26703.

A Numerical Investigation on the Effect of RF Coil Feed Variability on Global and Local Electromagnetic Field Exposure in Human Body Models at 64 MHz

Elena Lucano^{1,2}, Micaela Liberti², Tom Lloyd³, Francesca Apollonio², Steve Wedan³, Wolfgang Kainz¹, and Leonardo M. Angelone^{1,*}

¹US Food and Drug Administration, Center for Devices and Radiological Health, Silver Spring, Maryland, USA

²Universita degli Studi di Roma La Sapienza, Department of Information Engineering, Electronics, Telecommunications, Roma, Italy

³Imricor Medical Systems, Burnsville, Minnesota, USA

Abstract

Purpose—This study aims to investigate how the positions of the feeding sources of the transmit radiofrequency (RF) coil, field orientation direction with respect to the patient, and patient dimensions affect the global and local electromagnetic exposure in human body models.

Methods—Three RF coil models were implemented, namely a specific two-source (S2) feed and two multisource feed configurations: generic 32-source (G32) and hybrid 16-source (H16). Thirty-two feeding conditions were studied for the S2, whereas two were studied for the G32 and H16. The study was performed using five human body models. Additionally, for two of the body models, the case of a partially implanted lead was evaluated.

Results—The results showed an overall variation due to coil feeding conditions of the whole-body specific absorption rate (SAR) of less than 20%, but deviations up to 98% of the magnitude of the electric field tangential to a possible lead path. For the analysis with the partially implanted lead, a variation of local SAR at the tip of the lead of up to 60% was observed with respect to feed position and field orientation direction.

Conclusion—The results of this study suggest that specific information about feed position and field orientation direction must be considered for an accurate evaluation of patient exposure.

Keywords

SAR; finite difference time domain (FDTD); virtual population; interventional MRI; partially implanted leads

*Correspondence to: Leonardo Angelone, U.S. Food and Drug Administration, Center for Devices and Radiological Health Office of Science and Engineering Laboratories, Division of Biomedical Physics, 10903 New Hampshire Avenue, Bldg. 62, Rm. 1109, Silver Spring, MD 20993. Leonardo.angelone@fda.hhs.gov.

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 implied endorsement of such products by the U.S. Department of Health and Human Services.

SUPPORTING INFORMATION

Additional supporting information may be found in the online version of this article.

INTRODUCTION

MRI is a widely used diagnostic imaging technique that employs RF waves at Larmor frequencies to produce images. The exposure of the patient to the RF field during MRI is limited by specific guidelines (1). In commercial MRI systems, the exact feeding conditions of the RF coil (eg, feed position, field orientation direction) with respect to the patient often are unknown. It is expected that variation of RF coil-feeding conditions may result in variation of RF exposure in the body and the related possible RF-induced heating of tissue (2).

RF-induced heating may worsen in the presence of electrically conductive devices that are fully or partially implanted, or in contact with the patient due to the *antenna effect* of the device (3–6). Computational modeling increasingly has been used to complement experimental results to evaluate RF energy absorption near electrically conductive devices. The Technical Specification (TS) ISO 10974 (3) describes a methodology to computationally assess the exposure of devices fully implanted in the body. In particular, the document states how the study of a system in the meter range (ie, the RF coil), including an object in the μm range (ie, implant lead), implies a high computational cost when using computational approaches based on finite difference time domain (FDTD) method. An interesting solution included in the TS (ie, tier 3 approach) is the use of the so-called *transfer function*, first proposed by Park et al. (4). This approach consists of decoupling the problem of the lead sensitivity (ie, transfer function) with respect to the electric field incident to the lead from the effective exposure level. The power deposited in the tissue surrounding the lead tip can be computed by multiplying the transfer function of the specific lead with the incident electric field tangential to the lead. The tier 3 approach has been used by several studies (6–10); however, all are limited to fully implanted devices.

Currently, there is no standardized methodology to assess RF exposure in patients with partially implanted devices during MRI. MRI systems have been used beyond diagnostics to guide or track devices for real-time interventional procedures (interventional MRI (iMRI)) (11–13). When used to monitor catheter insertion inside the patient, such as in cardiac ablation or diagnostic catheterization, iMRI requires the presence of partially implanted leads. Evaluation of RF exposure in patients with partially implanted leads is further complicated because a portion of the catheter is outside of the body, where the electric fields may be high in magnitude and strongly influenced by coil design and coil-feeding conditions (14).

This study aims to provide a proof of concept on how the coil model feed variability may affect human body model exposure. Variability of global and local exposure was studied for five human body models landmarked at the heart inside the RF coil. Additionally, for two of the five models, the local exposure with a model of a lead partially implanted inside the body was evaluated. Three RF coil models with different feeding conditions were implemented, including a two-source feed (S2) and two multi-source feed configurations (G32 and H16). Variability of feeding conditions was evaluated with respect to field

orientation direction for all three of the coil models and with respect to feed source position for the S2 model only.

METHODS

Computational Models of RF Transmit Coils

Electromagnetic (EM) field simulations were implemented using the commercially available FDTD software Sim4Life V2.2 (ZurichMedTech, Zurich, Switzerland). The numerical CAD model of a high-pass RF transmit body coil at 64 MHz was implemented based on the MITS1.5 system available at the U.S. Food and Drug Administration Laboratories, as previously described in Lucano et al. (14) and briefly recalled here. The coil is composed of 16 rectangular strips (rungs) connected at each end by two end-rings and shielded by a 16-sided regular polygonal enclosure (Fig. 1a). Both the coil and the shield were modeled as perfect electric conductors (PEC).

Simulations were implemented following three modeling approaches (Figs. 1b–1d) similar to those reported in (14). The first model (S2) consisted on a two-source coil driven in quadrature mode (ie, equal amplitude with a 90° phase shift between each source excitation) with additional lossy components distributed in the coil end-rings. The electrical implementation of the S2 model used lossy components consisting in a resistor and a capacitor in parallel (ie, $C = 70$ pF and $R = 1$ k Ω). The second (G32) and third (H16) coil models were implemented with a multisource feed configuration. For all models, the feeding sources were composed of an ideal voltage source with a resistor of 50 Ω in series. The G32 used 32 50 Ω voltage sources distributed along the coil end-rings without any additional lossy components (15). The H16 model was based on a hybrid approach using 16 50 Ω voltage sources distributed within the middle of the rungs, and it included the same lossy components as the S2 model (16). The H16 model used in this study, implemented with voltage sources, was verified against the implementation with current sources. The maximum variability in terms of electric-field and magnetic-field magnitude and local SAR was less than 3%.

Computational Models of the Human Body

Five anatomically precise human body models, all part of the virtual population ViP 3.0 (17,18), were used to model a broad range of model variability. The models included two females (Ella and Billie) and three males (Glenn, Duke, and Fats), age range from 11 years old (Billie) to 84 years old (Glenn), and body mass index range from 15.3 (Billie) to 36.1 kg/m² (Fats). The position of each human body model within the RF coil was off-centered, with the model's heart at the isocenter of the coil (ie, heart landmark) in line with heart imaging during iMRI (Fig. 2a). The physical properties (ie, electrical conductivity, electrical permittivity, and mass density) of the anatomical structures of each model were based on existing literature for the specific frequency used (ie, 64 MHz) (19).

Coil Feeding Condition Variability

The human body model exposure to the EM fields was studied with respect to coil excitation variability. With the G32 and H16 coil models, the field exposure variability was studied with respect to field orientation direction:

$$\Delta\varphi: CW \wedge CCW \quad [1]$$

Specifically, the variability of φ was imposed with respect to the human body model by keeping the same current profile along the coil (ie, same phase difference between the feeding sources) and loading the coil with the body model either headfirst for clockwise (CW) orientation or feetfirst for counterclockwise (CCW) orientation. The fixed current orientation with respect to the chosen coordinate system allowed obtaining a B_1^+ field component always higher than the B_1^- . Conversely, if the φ had been obtained by switching the orientation of flowing currents inside the coil, the z -axis orientation of the coordinate system should have been reversed accordingly to ideal static field (B_0) to allow the prominence of B_1^+ over B_1^- .

With the S2 coil model, a total of 32 feeding conditions were defined by varying the field orientation direction, as in Equation [1], as well as the feed position within the coil (ie, 16 positions, 8 for each ring) (Fig. 2b), as follows:

$$\Delta\pi: \begin{cases} \alpha \in 22.5^\circ:45^\circ:337.5^\circ \\ ring: Head \wedge Feet \end{cases} \quad [2]$$

where α identifies the central angle for each pair and π indicates position variability (Fig. 2b), implemented by rotating the two-feed pair of 45° all around each of the two end-rings of the coil.

EM field propagation for the G32 and H16 coil models was implemented by simulating the current resonance profiles along the rungs of the coil. This was done by exciting each feeding source with a 64-MHz sinusoidal signal following a well-defined phase shift profile around the coil, as reported in (14). The same approach was applied to both cases without and with the lead.

Conversely, EM field propagation for the S2 coil model was numerically computed following two different approaches, one for the case without lead and one for the case with the lead. For simulations without the lead, each feeding source of the pair was excited by a broadband signal (20) from 0 to 140 MHz, with the peak at 64 MHz. For field extraction, results were postprocessed and combined, imposing the same signal amplitude and a phase shift of 90° at the two feeds. This procedure allowed researchers to implement a human body model exposures database in which results could be postprocessed anytime, imposing variable amplitude and a phase shift at the two sources. This process performed using a feature embedded within the Sim4Life software (ZurichMedTech). When the lead was

included in the model, the two feeds of each pair were simultaneously excited by a sinusoidal signal at 64 MHz, with the two signals at the same amplitude but 90° out of phase.

All simulation results were normalized to an average of Root Mean Square (RMS) of B_1^+ field ($B_{1\text{RMS}}^+$) of 3 μ Tesla (T) within the human body model under test in the central axial slice (Fig. 2b). This normalization was chosen according to the value suggested by the International Electrotechnical Commission (1). For the specific models and geometry selected in the study, such normalization corresponded to a range of power delivered by the coil to the load, at the net of all coil losses, between 18.1 watts (W) (Billie) and 108.5 W (Fats).

Quantities of Interest

The quantity of interest for evaluating global exposure of the human body model was the whole-body average SAR (WbSAR). The WbSAR variability was analyzed with respect to π (S2 only) and ϕ . The analysis was performed with the five human body models under test, defining an intersubject variability. This allowed to characterize the level of exposure with respect to the human body model when using a normalization based on the $B_{1\text{RMS}}^+$.

The quantities of interest for evaluating local exposure were the local peak of the SAR averaged over 10 g of tissue (10gSAR) and the electric field tangential to a predefined trajectory (E_{tan}) (3). The peak 10gSAR was evaluated for the five human body models, and the variability was analyzed with respect to π (S2 only) and ϕ . Both magnitude and phase of the E_{tan} were extracted (21), and for each model the selected trajectory was both inside (P_0 to P_1) and outside the body (P_1 to P_2) (Fig. 2c). The section inside the body was drawn along the vein path from the inferior vena cava down to the left femoral vein, and out from the vein at the groin. The section outside was drawn along the legs to mimic a possible catheter path that was the same for all models. Figure 2d reports the trajectories for the five human body models used. Variations of extraction trajectory were due to differences in body morphologies, namely: 1) length of the trajectory section inside the body, and consequently 2) P_1 location, which was at different coordinates with respect to the coil. The extraction of E_{tan} magnitude ($\|E_{\text{tan}}\|$) and phase was performed using the IMSAFE tool embedded inside the Sim4Life software (ZurichMedTech).

Additional analysis, including the model of a partially implanted lead, was performed with two of the human body models (ie, Glenn and Duke). The exposure was studied for the power absorbed in the tissue near the lead tip (SAR_{tip}). The SAR_{tip} was analyzed in relation to the local quantities: unaveraged SAR (uSAR), 0.1g average SAR (0.1gSAR), 1g average SAR (1gSAR), and 10gSAR. The computation of the averages SAR was performed using the internal algorithm embedded in the Sim4Life software (ZurichMedTech) at the same location for all averages.

Computational Model of Partially Implanted Lead

The lead consisted of a 1.19 ± 0.01 -m long partially insulated wire (ie, 2-mm diameter wire covered by a 0.5-mm insulation) with a 7-mm bare tip. The lead was inserted in the

body at the groin region, terminating in all models 40 mm inside the body by the left femoral vein. The position of the lead was in line with the initial stage of a catheterization ($P_0^* - P_2$ in Figure 2d). Within the numerical environment, the wire was modeled as a PEC and the insulator as Teflon ($\sigma = 4.6 \text{ e-}3 \text{ S/m}$, $\epsilon = 2.08$). Details about the numerical mesh used for the coil and lead are reported in the Supporting Information.

RESULTS

Global Exposure

The simulation results with all the models showed WbSAR values lower than 2 W/kg with the selected normalization, within the safety limit in the normal operating mode (1). Figure 3 reports the WbSAR values observed for each human body model under test with respect to: 1) π and ϕ for the S2 coil (see radar plot), and 2) ϕ for the G32 and H16 models (see Table in Fig. 3). The radar plot reports the WbSAR relative to the specific feeding source pair identified by polar angle of the plot (ie, internal angle α of the two feeds) (Fig. 1b). Inside each radar plot, the human body model under test is reported in scale with the coil profile identified by the external bold circle in black. The representation of the results based on radar plot allows for an immediate estimation of π variability; the closer the distribution to a circular shape, the less π variability is present.

The intrasubject variability of WbSAR ranged between 12.7% and 18.7% with respect to π and between 1% to 10% with respect to ϕ . Table 1 shows the maximum percentage WbSAR variability observed within each body model with respect to π and ϕ . The highest intrasubject variability was observed within Glenn both with respect to π and ϕ . Additionally, the intersubject variability was up to 57%, with maximum WbSAR obtained with Glenn and minimum WbSAR obtained with Billie.

With the G32 and the H16 models, WbSAR values were similar to those obtained with S2 model (Fig. 3, see Table), but with a smaller standard deviation due to variability only related to ϕ by design. When comparing the results obtained with G32 versus H16, there was less than 2% variability.

Local Exposure

Results showed that the location of the 10gSAR peak varied for each body model (eg, in the neck with Duke, in the lateral middle chest with Glenn). Moreover, the specific location of the peak 10gSAR for each model was independent with respect to π , whereas with ϕ it was specular with respect to the longitudinal axis of the body (z-axis). Peak 10gSAR values ranged from 6.6 W/kg (in Billie) to 15.4 W/kg (in Glenn and Fats). The intrasubject variability ranged between 30.4% (in Duke) and 47.1% (in Ella) with respect to π , and ranged between 0.2% (in Duke) to 39.6% (in Glenn) with respect to ϕ . See the Supporting Information (Supporting Fig. S2) for the full range of values.

The variability of $\| \mathbf{E}_{\tan} \|$ and WbSAR was evaluated with respect to coil feed conditions with the S2, G32, and H16 coil models (Fig. 4) and the five body models. See the Supporting Information (Supporting Fig. S3) for the profiles of the $\mathbf{E}_{\tan}$ phase. As shown in Figure 4, results section by section for the S2 showed that:

1. Inside the body (P_0 to P_1 , gray area in Fig. 4), the highest $\| \mathbf{E}_{\tan} \|$ dynamic range was observed in Duke, with a peak of 67.7 volts per meter (V/m) and a nadir of 0.4 V/m. In Glenn, Duke, and Fats, the results in CW polarization showed higher $\| \mathbf{E}_{\tan} \|$ with respect to the CCW exposure for most of the trajectory. Conversely, the opposite was observed in Ella and Billie, indicating that neither of the two polarizations can be assumed as the worst case for $\| \mathbf{E}_{\tan} \|$ over the five human body models considered.
2. For the interface groin/air, the highest $\| \mathbf{E}_{\tan} \|$ peaks were observed with Glenn and Billie, namely 183 V/m and 166 V/m, respectively. This is related to the specific location of the P_1 , close to the coil end-ring (ie, $z = -325$ mm) (Fig. 2d). Conversely, Fats showed the lowest peak of 77 V/m.
3. Outside the body (P_1 to P_2), for the case defined in the study, the trajectory was always outside the coil. At the beginning of the trajectory, Duke, Fats, and Ella showed well-defined double peaks of electric field. This was related to the specific trajectory bending profile closed for the extraction. Along the trajectory, isolated peaks along the profile were related to the vicinity of the extraction trajectory to the body of the model. For each model, the highest peak was found at the end of the trajectory (P_2), near the feet. This was due to the antenna effect of the body itself, with induced currents along the body that generated a high peak of the E_z component in the proximity of the feet. Peaks values at the feet, even up to 1.32 m away from the coil (ie, in Fats), were of the same order of magnitude of the $\| \mathbf{E}_{\tan} \|$ inside the body. The intersubject analysis showed the highest peak at the feet (ie, 57 V/m) for Glenn.

Table 1 reports the maximum percentage of $\| \mathbf{E}_{\tan} \|$ variability observed within each body model with respect to π and φ . The results show a change of up to 93% with respect to π and up to 98% with respect to φ .

Results obtained with the G32 and H16 coil models corresponded with the one obtained for S2 but averaged with respect to the π variability for the specific field-orientation direction studied (ie, CW or CCW). Profiles are reported in Figure 4. As for the WbSAR, when comparing the results obtained with G32 versus H16, there was less than 2% variability.

Human Body Model With Partially Implanted Lead

The variability of SAR_{tip} showed changes up to 51% with Glenn and up to 60% with Duke with respect to φ (Table 1). Changes with respect to π were up to 50% for both models. The results of the uSAR, 0.1gSAR, 1gSAR, and 10gSAR showed all the same variability with respect to π and φ . This was expected because the local exposure is linearly proportional with respect to the specific mass used to average the quantity of interest. Hence, the radar plots in Figure 5 report profiles valid for all four quantities but normalized to the maximum value of the quantity of interest (in Table 2; values in *italics*). For each model, the dynamic range of results within the radar plot suggests the variability for different feed locations and field orientation direction. Specifically, the dynamic range was 0.56 for Glenn and 0.64 for Duke; with a maximum value equal to 1 obtained with head CW in 157.5°

position in both models, whereas the minimum value equal to 0.44 and 0.36 for Glenn and Duke, respectively, obtained with the feet CCW in 157.5° position.

The highest SAR_{tip} (Table 2; values in italics) with the S2 model was obtained with the head-CW. Results with the multisource models (G32, H16) were lower than the maximum obtained with the S2 model (eg, Glenn, uSAR S2/head CW vs. G32/CW 28%), but they were within the π variability range (eg, Glenn, uSAR S2/CW from 1,189 W/kg to 2,354 W/kg vs. G32/CW equal to 1,700 W/kg). Notably, whereas only one distribution relative to one coil feeding condition is shown, the same distributions—with different absolute values—were obtained when varying π and ϕ .

DISCUSSION

The main contribution of this study is to provide a proof of concept on the effect of different feeding conditions of an MRI RF coil at 64 MHz on global and local exposure in human body models. Five different human body models were considered in the analysis to analyze intersubject variability (16). For each human body model, variability with feeding conditions was achieved generating the EM field exposure with three implementation approaches of the RF body coil model (ie, S2, H16, and G32 similar to the one used in (14)). The selected quantities of interest were the variability of WbSAR, local 10gSAR, as well as E_{tan} along a predefined path both inside and outside the body.

WbSAR results and intersubject variability were in line with a previous numerical study (16). Such an intersubject variability was mainly due to the applied normalization with $B_{1\text{RMS}}^+$. The WbSAR is affected by factors such as body shape, muscle/fat ratio in the body, and total body mass due to the specific normalization used (ie, $B_{1\text{RMS}}^+$). The B_1^+ homogeneity in a human body model decreases getting closer to the coil rungs; thus, larger sections of the body result in lower averages and higher normalization factors. The feed position did not affect the specific position of the peak 10gSAR value but only its absolute value. Conversely, the position was affected by ϕ due to the resulting opposite orientation of the induced currents within the body.

The $\|E_{tan}\|$ profiles along the selected trajectory also were affected by the specific body model, feed position, and field orientation direction. $\|E_{tan}\|$ results were characterized by peaks (inside the body, interface groin/air, feet) and nadirs. All five models showed a jagged $\|E_{tan}\|$ profile for the trajectory section inside the body, a high $\|E_{tan}\|$ peak at the interface groin/air, and a smoother profile for the trajectory section outside the body. Along the section inside the body (P_0 – P_1), the E_{tan} was affected by all three field components (ie, E_x , E_y , and E_z) because the trajectory was not along a preferred axis. Conversely, along the section outside the body (P_1 – P_2) (with the exception of the first bended section) the trajectory was user-defined along the z -axis (Fig. 2c); thus, the E_{tan} profile only was affected by the E_z component. Peak values at the groin/air interface were affected by the specific location with respect to the coil rather than by the overall exposure. High values of $\|E_{tan}\|$ were observed with both Glenn and Billie (the point P_1 was close to the coil ring in both models), although the WbSAR with Glenn was twice as much as the WbSAR with Billie. Conversely, the $\|E_{tan}\|$ peak in Fats was 54% lower with respect to Billie, even if the

WbSAR of Fats was 55% higher than the one of Billie. This result was due to the location of P_1 , which in Fats was 240 mm farther from the coil compared to the position in Billie.

Compared to the G32 and H16 coil models, the feeding position variability was the additional degree of freedom that allowed the S2 model to generate a broader range of exposure conditions. Consequently, the G32 and H16 coil models were able to generate results in line with the S2 model for the case studied, but possible underestimations and defects with respect of the worst-case exposure may need to be taken into account.

For the S2 model, the combination of the asymmetric loading conditions due to the selected landmark, both along the coronal (z -axis) and axial (y -axis) directions, as well as the π affected the coupling between the human body model and the coil (see Supporting Information). The WbSAR values reported in Figure 3 also are related to such variability, and their distribution followed similar symmetry (eg, similar WbSAR values) with opposite feed locations, 22.5 and 202.5, with similar coupling conditions. For the specific conditions analyzed in this study, the range of variability of the resonant frequency across each human body model and the 32 feeding conditions was less than 0.3% (see Supporting Information). Simulated near-field and local exposure, however, may be also affected by tuning and matching conditions (eg, inclusion of feeding subcircuit) and coil losses (eg, reflected power, lossy components), as reported in (9).

The coupling of conductive leads with the tangential component of the electric field is responsible for the induced heating at the tip of a lead. When the transfer function of the lead (3) is unknown, a possible approach is the use of a simplified transfer function to qualitatively determine a variability of power deposition with respect to different exposures (10). Herein, simplified transfer function with constant magnitude equal to 1, and two phase slopes (ie, linear to -5 rad, and linear to -10 rad) were implemented as an example. The results showing variations of computed power deposition due to the $E_{\tan}$ results can be found in the Supporting Information (Supporting Fig. S4 for linear to -5 rad phase; Supporting Fig. S5 for linear to -10 rad phase).

Additional analysis was performed with a model of a lead partially implanted in a human body model. Technical specifications for safety assessment of fully implanted medical devices in MRI have been presented (3) and studied in literature (5,6,22), whereas no standardized methodology exists for partially implanted devices. Simulations with the lead were implemented using two of the five human body models as a tradeoff between the need to evaluate intersubject variability and the computational cost of the simulations. The selection of the models was based on the high use in literature (ie, Duke) and the interest to a model of an older patient (ie, Glenn), likely the best patient candidate for an MRI interventional catheterization procedure. The quantity of interest for this analysis was the local SAR in the structure surrounding the tip of the lead.

As per antenna theory (4,5,23), given a specific lead geometry the local SAR depends on the $E_{\tan}$ profile, which in turns depends on the trajectory of the lead with respect to the incident field. In this study, the trajectory inside the body was constrained by the morphology of the human body model, following typical catheter-insertion procedures (ie, through femoral

vein). Conversely, the trajectory outside was chosen to follow a straight path as an example. It is expected that results would vary with variable trajectories outside the body, different body postures, or landmark positions. Additionally, as a verification of the computational framework, the results of SAR distribution near the tip were evaluated. Although the lead is an elongated structure, the tip of the lead can be considered electrically small. As such, although the absolute value of peak SAR depends on the lead exposure, the spatial distribution of SAR around the tip of the lead depends only on the tip shape, which is in line with previous literature (4,5), as shown in Figure 6. Notably, small variations of the SAR spatial distribution are visible in the figure due to differences in the lead tip modeling voxelization.

Limitations

The results presented are specific to the models used and cannot be directly extrapolated for a clinical environment. As following development of this study, experimental validation of the results will be performed using a phantom to validate the results with respect to the coil feeding condition variability. Different E_{tan} extraction paths, lead trajectory, or penetration depths would result in different range of results variability with respect to π and ϕ . Finally, to be able to apply the results obtained with the partially implanted lead to assess RF-induced thermal damage, additional thermal analysis and full uncertainty evaluation are needed. Uncertainty evaluation of the power deposition at the lead tip should also include the effect of the FDTD staircasing and a thorough convergence analysis with respect to the FDTD grid used (24).

CONCLUSION

The results of this study indicate that the specific information about feed position and field orientation direction with respect to the human body model must be considered when evaluating local SAR in the presence of a conductive elongated structure within, or in contact with, the body of the patient. For the specific models used, results showed a variation of less than 20% for WbSAR, up to 47.1% for the local 10gSAR, and up to 98% of the incident electric field magnitude along the single trajectory analyzed. The analysis with the partially implanted lead showed changes up to 60% of peak SAR at the lead tip with respect to field orientation direction. Intersubject analysis performed with five human body models showed up to 58% variability in WbSAR. The results of this study should not be taken as a safety guide and are limited to the specific geometries evaluated; additional work is needed to validate the numerical outcomes— and to include multiple lead paths and penetration depths, body postures, and landmark positions.

Supplementary Material

Refer to Web version on PubMed Central for supplementary material.

Acknowledgments

This study was in part funded by a Cooperative Research and Development Agreement (CRADA) between Imricor Medical Systems and the U.S. Food and Drug Administration. The authors would like to thank the following

colleagues: Brian Beard, Daniel Hammer, Maria Iacono, Victor Krauthamer, Sunder Rajan, Amir Razjouyan, Devashish Shrivastava.

References

1. IEC 60601-2-33 ed3. Particular requirements for the basic safety and essential performance of magnetic resonance equipment for medical diagnosis. 2010
2. Yeung CJ, Susil RC, Atalar E. RF heating due to conductive wires during MRI depends on the phase distribution of the transmit field. *Magn Reson Med*. 2002; 48:1096–1098. [PubMed: 12465125]
3. ISO, TS 10974 Assessment of the safety of magnetic resonance imaging for patients with an active implantable medical device. 2012
4. Park S-M, Kamondetdacha R, Nyenhuis JA. Calculation of MRI-induced heating of an implanted medical lead wire with an electric field transfer function. *J Magn Reson Imaging*. 2007; 26:1278–1285. [PubMed: 17969143]
5. Langman DA, Goldberg IB, Finn JP, Ennis DB. Pacemaker lead tip heating in abandoned and pacemaker-attached leads at 1.5 Tesla MRI. *J Magn Reson Imaging*. 2011; 33:426–431. [PubMed: 21274985]
6. Cabot E, Lloyd T, Christ A, Kainz W, Douglas M, Stenzel G, Wedan S, Kuster N. Evaluation of the RF heating of a generic deep brain stimulator exposed in 1.5 T magnetic resonance scanners. *Bioelectromagnetics*. 2013; 34:104–113. [PubMed: 23060256]
7. Zastrow, E., Capstick, M., Cabot, E., Kuster, N. IEEE 2014 International Symposium on Electromagnetic Compatibility. Tokyo, Japan: 2014. Piece-wise excitation system for the characterization of local RF-induced heating of AIMD during MR exposure; p. 241-244.
8. Li, D., Zheng, J., Liu, J., Chen, Y., Pan, C., Kainz, W. IEEE 2014 International Symposium on Electromagnetic Compatibility. Tokyo, Japan: 2014.
9. Kozlov M, Lucano E, Angelone LM. Effects of tuning conditions on near field of MRI transmit birdcage coil at 64 MHz. *Conf Proc IEEE Eng Med Biol Soc*. 2016; 2016:6242–6245. [PubMed: 28269678]
10. Mattei E, Lucano E, Censi F, Angelone LM, Calcagnini G. High dielectric material in MRI: numerical assessment of the reduction of the induced local power on implanted cardiac leads. *Conf Proc IEEE Eng Med Biol Soc*. 2016; 2016:2361–2364. [PubMed: 28324965]
11. Nitz WR, Oppelt A, Renz W, Manke C, Lenhart M, Link J. On the heating of linear conductive structures as guide wires and catheters in interventional MRI. *J Magn Reson Imaging*. 2001; 13:105–114. [PubMed: 11169811]
12. Jolesz FA. Interventional and intraoperative MRI: a general overview of the field. *J Magn Reson Imaging*. 1998; 8:3–7. [PubMed: 9500253]
13. Yeung, CJ., Susil, RC., Atalar, E. RF Safety of wires in interventional MRI : using a Safety Index. Conference Proceedings of the 23rd Annual International Conference of the IEEE Engineering in Medicine and Biology Society; Istanbul, Turkey. 2001; p. 2496-2498.
14. Lucano E, Liberti M, Mendoza G, Lloyd T, Iacono MI, Apollonio F, Wedan S, Kainz W, Angelone L. Assessing the electromagnetic fields generated by a radiofrequency MRI body coil at 64 MHz: defeaturing vs. accuracy. *IEEE Trans Biomed Eng*. 2016; 63:1591–1601. [PubMed: 26685220]
15. Liu W, Collins CM, Smith MB. Calculations of B1 distribution, specific energy absorption rate, and intrinsic signal-to-noise ratio for a body-size birdcage coil loaded with different human subjects at 64 and 128 MHz. *Appl Magn Reson*. 2005; 29:5–18. [PubMed: 23565039]
16. Murbach M, Neufeld E, Kainz W, Pruessmann KP, Kuster N. Whole-body and local RF absorption in human models as a function of anatomy and position within 1.5T MR body coil. *Magn Reson Med*. 2014; 71:839–845. [PubMed: 23440667]
17. Christ A, Kainz W, Hahn EG, et al. The virtual family—development of surface-based anatomical models of two adults and two children for dosimetric simulations. *Phys Med Biol*. 2010; 55:N23–N38. [PubMed: 20019402]
18. Gosselin M-C, Neufeld E, Moser H, Huber E, Farcito S, Gerber L, Jedensjö M, Hilber I, Di Gennaro F, Lloyd B. Development of a new generation of high-resolution anatomical models for

medical device evaluation: the Virtual Population 3.0. *Phys Med Biol*. 2014; 59:5287–5303. [PubMed: 25144615]

19. Gabriel S, Lau RW, Gabriel C. The dielectric properties of biological tissues: II. Measurements in the frequency range 10 Hz to 20 GHz. *Phys Med Biol*. 1996; 41:2251–2269. [PubMed: 8938025]
20. Murbach M, Neufeld E, Samaras T, Córcoles J, Robb FJ, Kainz W, Kuster N. Pregnant women models analyzed for RF exposure and temperature increase in 3T RF shimmed birdcages. *Magn Reson Med*. 2017; 77:2048–2056. [PubMed: 27174499]
21. Balanis, CA. *Antenna Theory: Analysis and Design*. Hoboken, NJ: John Wiley & Sons; 2012.
22. Mattei E, Triventi M, Calcagnini G, Censi F, Kainz W, Mendoza G, Bassen HI, Bartolini P. Complexity of MRI induced heating on metallic leads: experimental measurements of 374 configurations. *Biomed Eng Online*. 2008; 7:11. [PubMed: 18315869]
23. Guy AW. *Biophysics. Energy absorption and distribution*. AGARD Lect Ser. 1974; 78:1–14.
24. Cangellaris AC, Wright DB. Analysis of the numerical error caused by the stair-stepped approximation of a conducting boundary in FDTD simulations of electromagnetic phenomena. *IEEE Trans Antennas Propag*. 1991; 39:1518–1525.

FIG. 1.

Electrical characterization of the RF coil models. **(a)** 3-D view of the computational model. The computational RF body coil system was modeled to match the physical coil geometry: **(b)** S2, **(c)** G32, and **(d)** H16. RF, radiofrequency; G32, generic 32-source model; H16, hybrid 16-source model; S2, specific two-source model.

FIG. 2.

Numerical setup and data analysis definition. **(a)** Human body model Duke loaded in the RF coil landmarked at the heart, the black line shows the trajectory used for the E_{tan} extraction.

The $B_{1\text{ RMS}}^+$ at the central axial slice for the area occupied by the body is shown. All the numerical results were normalized based on the $B_{1\text{ RMS}}^+$ average equal to 3 μT in the selected area. Head and feet end-rings are defined. **(b)** S2 feed rotation within the coil end-ring where α defines the central angle of the two feeds at 90° used as 50 Ω voltage sources in the simulations. The field orientation direction (CW and CCW) also is shown. A total of 32 feeding conditions were studied with the specific two-source coil model. **(c)** Extraction line trajectory shown both inside (P_0 – P_1) and outside (P_1 – P_2) the body for the E_{tan} evaluation. **(d)** Extraction line trajectories for the five human body models used for the analysis. In Glenn and Duke, P_0^* indicates the point corresponding to the tip of the implant used in the models with partially implanted lead. CCW, counterclockwise; CW, clockwise; E_{tan} , electric

field tangential to a predefined trajectory; RF, radiofrequency; T, tesla; RMS, root mean square.

Author Manuscript

Author Manuscript

Author Manuscript

Author Manuscript

FIG. 3.

WbSAR results for the five human body models used for the analysis. The radar plots report the values with respect to π and ϕ for the S2 model. The bold external circle indicates the coil with each human body model inside plotted in scale. The table reports the WbSAR mean and standard deviations with respect to all the variables: π and ϕ for S2 and ϕ for G32 and H16. G32, generic 32-source model; H16, hybrid 16-source model; S2, specific two-source model; Std = standard deviation; W, watts; WbSAR, whole body average specific absorption rate.

FIG. 4.

$\| \mathbf{E}_{\text{tan}} \|$ profiles for each human body model studied along the extraction trajectory (Figs. 2c and 2d). The gray area of each plots identifies the extraction section inside the body. Results are shown as stripes (identified by two colored lines), including π and ϕ variability for the S2 model. Profiles obtained with G32 and H16 are plotted as black lines, with ϕ variability reported by solid (clockwise) and dotted (counterclockwise) lines. $\mathbf{E}_{\text{tan}}$, electric field tangential to a predefined trajectory; G32, generic 32-source model; H16, hybrid 16-source model; S2, specific two-source model.

FIG. 5.

Values of SAR_{tip} obtained for the different lead exposures (π and ϕ) with each of the two human body models selected for the analysis: Glenn (a) and Duke (b). Whereas the peak SAR depends on the lead exposure, the spatial distribution around the electrically small tip of the lead depends only on tip geometry. Thus, the profiles reported apply to either the uSAR, 0.1gSAR, 1gSAR, or the 10gSAR normalized to its maximum value (see Table 2; values in *bold and italic*). CCW, counterclockwise; CW, clockwise; gSAR, gram average SAR, specific absorption rate; SAR_{tip} , SAR near the lead tip; uSAR, unaveraged SAR; W, watts.

FIG. 6.

Map of uSAR around the tip of the partially implanted lead for one of the tested exposure conditions in Glenn and Duke. Values are reported as isosurfaces of SAR level with respect to the maximum value. The highest values were observed at the lead tip at the interface between bare tip and insulation and at the location where the lead exited the vein. SAR, specific absorption rate; uSAR, unaveraged SAR; W, watts.

Global and Local Exposure Percentage Change with Respect to π and φ (S2 Model Only). WbSAR and $\|\mathbf{E}_{\text{tan}}\|$ Along the Entire Trajectory (P0–P2) for the Five Human Models Studied. SAR_{tip} Peak Results are Reported Only for the Two Human Body Models Selected for the Case Study

Table 1

Maximum Variability with S2						
		Glenn	Duke	Fats	Ella	Billie
WbSAR	π	18.7%	16.6%	12.7%	18.0%	17.5%
	φ	9.5%	7.3%	2%	1.4%	0.8%
$\ \mathbf{E}_{\text{tan}}\ $	π	64%	92%	86%	93%	93%
	φ	93%	96%	93%	98%	98%
SAR _{tip}	π	49%	50%	–	–	–
	φ	51%	60%	–	–	–

$\mathbf{E}_{\text{tan}}$, electric field tangential to a predefined trajectory; SAR, specific absorption rate; SAR_{tip}, SAR near the lead tip; WbSAR, whole body average SAR.

Table 2

Maximum SAR_{tip} for All the 32 Feeding Conditions (π and ϕ) Defined in the S2 Coil Model and with Respect to ϕ for the G32 and H16 Coil Model.

Maximum SAR _{tip}										
Glenn					Duke					
	[W/kg]	uSAR	0.1 g	1g	10g	[W/kg]	uSAR	0.1 g	1g	10g
S2	Head CW	2354	285	81	12.4	S2	Head CW	2983	398	112
	Feet CW	2134	256	73	11.3		Feet CW	2557	345	97
	Head CCW	1914	213	62	9.7		Head CCW	2152	275	76
G32	Feet CCW	2028	221	65	10.2		Feet CCW	2170	282	78
	CW	1700	205	58.2	9.0	G32	CW	2279	304	85.5
	CCW	1412	152	44.9	7.2		CCW	1543	196	54.2
H16	CW	1748	211	60.0	9.3	H16	CW	2271	302	84.9
	CCW	1351	146	43.0	6.9		CCW	1523	194	53.6

CCW, counterclockwise; CW, clockwise; G32, generic 32-source model; H16, hybrid 16-source model; S2, specific two-source model; SAR, specific absorption rate; uSAR, unaveraged SAR; W, watts.