Using Scaling Approach to Estimate MRI RF Field Induced Heating for Small Medical Implant

Dawei Li, Jianfeng Zheng,Yan Liu, Ji Chen
Dept. of Electrical and Computer Engineering
University of Houston
Houston, USA
{dli15, jzheng3, jchen18}@central.uh.edu

Changwan Pan
Micro-Tech Co., Ltd.
Nanjing, P. R. China

Wolfgang Kainz
U.S. Food and Drug Administration
Rockville, MD, USA
wolfgang.kainz@fda.hhs.gov

***Abstract*—In this paper, a quick and efficient approach is proposed to estimate the in-vivo RF induced heating of small medical implants under MRI procedure. The method is based on scaling the in-vivo incident electric fields. Numerical and experimental studies demonstrate the efficiency and accuracy of the proposed method.**

***Keywords*—RF induced heating; MRI safety; scaling approach; medical implants;**

I. Introduction

Metallic implantable devices are widely used for clinical therapies to support a damaged biological tissue or structure. When patients with implantable devices undergo magnetic resonance image (MRI) for diagnosis, one of the main risks is the induced heating due to the interaction between medical implants and MRI radiofrequency (RF) coils [1]. Numerous researches have been conducted to study the RF heating for various medical implants. For example, Sommer, et al. [2] studied the heating effects of certain pacemaker leads and founded that, depending on different electrodes, temperature rise vary from 0.1°C to 23.5°C. Kainz, et al. [3] reported a RF compatible neurological pulse generators produce a maximum of 2.1°C temperature increases at the lead tip.

Most medical implants belong to one of the three categories. The first category is the active stimulator, for example, cardiac pacemaker and neuro-stimulation systems. The second is the orthopedic fixations which are used to anchor fractured bones while they are healing. The third type is bioactive devices, such as drug-eluting stents to prevent the fibrosis of the tissue. Various methods have been proposed to evaluate the MRI RF coil induced heating effect of these medical implants [4-6]. However, using electromagnetic modeling alone maybe ineffective and computational expensive when the medical devices have small features since fine meshes are needed to model the small features of implant devices. Consequently the required computation costs will greatly increase with the mesh resolution and the number of unknowns. When it comes to the situation such as medal stent, sometimes it maybe computational prohibitive due to the required computer memory and the required CPU time, especially if one wants to estimate the temperature rises of such stents inside human subjects via modeling approach. For example, the mesh resolution for human body lower than 2 mm exceeds the capability of most computation-accelerated GPU, while in order to reduce error of geometry modeling for small devices, the mesh resolution is around 0.2 mm depending on the structure. Essentially, the problem becomes to a multi-scale problem due to the different size between human body and small device. In order to solve this multi-scale problem, the proposed method models the medical implant and human separately, and analyzes the two parts in different scales. The in-vivo heating pattern is then evaluated by integrating the parameters obtained from two different scaled simulations.

The paper is organized in the following. Firstly, the heating mechanism for RF heating is analyzed. Secondly, a scaling approach is proposed by taking the advantage of the feature of small structure to simplify the evaluation. Then a 10 cm rod and a medical stent are used as examples to validate the method under ASTM phantom. After validation, a geometrical dependent heating factor for stent is developed. Finally, the in-vivo heating diagrams are plotted according to the simulated incident field in anatomic accurate human body model from virtual family.

II. Methodology

A. RF induced heating

The RF heating of the medical implants under MRI environments is considered as one of the limiting factor whether patients with medical implants can be scanned. This is a very complex process since such induced heating is related with RF coil type, human body type, human subject loading position, the geometry of implantable devices, constructing material of medical implants, and etc. In the field of electromagnetics, the energy dissipated in the lossy biological tissues is described via the Specific Absorption Rate (SAR). The SAR is defined as

$$SAR(r) = \frac{\sigma}{2\rho} E^2(r) \tag{1}$$

where E is the total electric field, and σ , ρ are the conductivity and density of biological tissue respectively. The temperature rise in human body caused by the dissipated energy in tissues can be demonstrated by the total SAR in the volume according to the bio heat equation. The total electric field equals to the incident field from the RF coil and scattering field induced by the medical implants. The heating effect around device tip is a localized phenomenon and the temperature rise around device tip for the 15 minutes MRI scan round is closely related to the SAR values near the device tips.

The RF induced heating near implantable devices can be simplified as a scattering problem of implanted structures buried in lossy medium under particular sources. The lossy material is the gel inside ASTM phantom with conductivity around 0.47 S/m. The incident field radiating from the RF coil penetrates into ASTM phantom and interacts with the metallic devices. During such interactions, a surface current is induced on the device to generate a scattering field in order to satisfy the boundary condition. It is observed that the largest total field is observed around the device tip where charges accumulate. To solve the problem is to find the near field around the device tip.

The total field is related to the incident field by this simple integral.

$$E_{tip}^{total} = \int_S \underline{E}^{inc} \cdot \underline{T}\, dS \tag{2}$$

where $\underline{E}^{inc}$ is the incident field along the device and the $\underline{T}$ is a complex function (often time refers as the transfer function) related to the device geometry.

B. Quick Estimation Scheme

However, obtaining the complex transfer function of complex geometry requires significant efforts either in measurement or electromagnetic modeling. In some cases, it can be computational prohibitive due to the required computational memory and CPU time. In addition, the obtained transfer function can only be represented in a complex 3D data format and maybe not be readily used for in-vivo integration. Therefore, alternative approach is desirable.

Assuming that the device is not electrically large, the incident field will then not have significant variation over the entire device body. Under this circumstance, the incident field can be factored outside the integral of (2). This is valid as long as the dimensions of the implants is small compared with wavelength. In addition, the phase variation of the incident field is very small as well. Therefore, such almost constant incident field can be taken out of the integral. After factoring out the incident electric field, the integral becomes

$$E_{tip}^{total} = \underline{E}^{inc} \cdot \int_S \underline{T}(r)\, dS \tag{3}$$

In this case, the incident fields are decoupled with the transfer function mentioned before. Because the temperature rise is roughly linearly proportional to the SAR after 15 minutes MRI scan, we get the equation below.

$$\Delta T \approx C_1 \sigma \left| \underline{E}^{inc} \cdot \int_S \underline{T}(r)\, dS \right|^2 \approx C \left| \int_S \underline{T}(r)\, dS \right|^2 \left| \underline{E}^{inc} \right|^2 \tag{4}$$

Instead of evaluating this complex transfer function on the surface of the implants, the integral of transfer function alone is sufficient for us to determine the total field around device tip. From the analysis above, the original problem is divided into two steps. First, the human model is loaded in the RF coil to obtain the incident field at all implantable locations. The mesh resolution for human body is 2 mm. Then, (4) is evaluated experimentally by placing the medical implants inside the ASTM phantom and measure the temperature rises at the location where the maximum heating is expected. In addition, the incident electric field at the ASTM phantom location where the device is placed is evaluated either via experimental measurement or numerical modeling. After the total electric field is obtained, a heating factor is determined by the ratio of the measured temperature rise to the intensity of incident electric field onto the medical device.

III. Numerical Modeling and Experiments

Measurement and modeling are used in the following to demonstrate the efficiency of our approach.

A. Cylindrical Rod

A 10 cm length rod with a diameter of 1cm is placed inside the ASTM phantom. The ASTM phantom is placed in the center of RF coil as described in [8].Twelve different locations inside ASTM phantom are chosen to represent different incident field distribution. Among them, eight placements are parallel to the long side wall, while 4 placements are parallel to the short side wall. A graph illustrates the position of the testing cylindrical rod with the number below indicating the simulation index. The cylindrical rod here is modeled as perfect electric conductor (PEC). Commercial finite-difference time-domain based solver, SEMCAD was used to simulate the field distribution around the cylindrical tip.

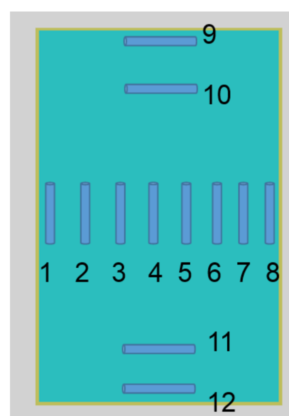

Fig. 1 10 cm Rod positioning inside ASTM phantom

Table 1 shows the comparison of electric field around rod tip between full wave simulation and scaling approach proposed from the analysis above. Since the electric field is singular near the device tip, the volume-averaged E field defined by ICNIRP-2010 is used for comparison. Because the gel inside the phantom is a lossy material, the incident electric field is larger near the side wall. The ratio of largest electric field to the smallest electric field is about 3. It is expected that larger incident field along the phantom wall gives a larger total field with cylindrical rod present. This corresponds to two peaks at simulation index 1 and 8 shown in the table.

TABLE 1 TABLE OF ELECTRIC FIELD AND TEMPERATURE RISE

Simulation Index	***Comparison of Simulation and Estimation***			
	Estimated Electric Field(V/m)	***Simulated Electric Field(V/m)***	***Estimated Temperature Rise(°C)***	***Simulated Temperature Rise(°C)***
1	1430	1430	9.24	9.49
2	1210	1170	6.61	6.63
3	810	743	2.96	3.07
4	360	321	0.58	0.49
5	372	331	0.63	0.52
6	820	758	3.03	3.23
7	1220	1190	6.73	6.72
8	1412	1402	9.01	9.09
9	800	886	2.89	3.89
10	586	636	1.55	1.87
11	586	635	1.55	1.86
12	800	886	2.89	3.18

The comparison of temperature rise around rod tip between direct simulation and scaling approach is also given in the table above. The estimated temperature rise is obtained from the intensity of the electric field using a linear constant. From the table, this linear factor is about 0.0283 °C / (kV / m)^2. This means 100 V / m strength of electric field in gel induces a temperature rise of 2.830°C after 15 minutes MRI scan. Direct temperature rise is obtained by simulation using SEMCAD Pennes solver [7]. Again good agreement is observed between the direct simulation and scaling estimation method.

B. *Medical Implants(Metallic Stent)*

The second example is a 10 cm long medical stent. Using electromagnetic simulations, it is determined that the maximum SAR will be located at the end of stent. Because the accurate simulation of the transfer function costs a very longer time to converge, only direct measurements were conducted in this case. The stent was put at 6 different locations parallel to the long side wall with 6 cm spatial resolution. Thermal probe was put in the worst heating location to measure the temperature rise. Then the incident fields inside ASTM phantom at other possible placement locations, as shown in Fig. 3, are extracted. Probe was also placed in the symmetric location inside ASTM phantom to capture background temperature rise.

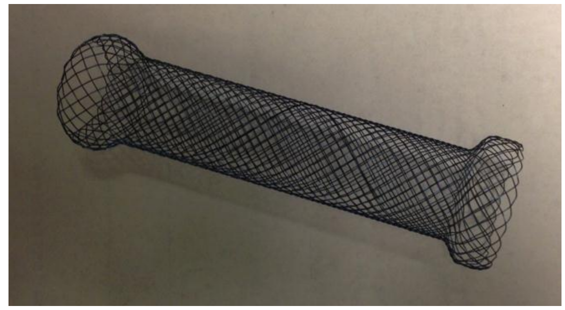

Fig. 2 10 cm length medical implantable stent

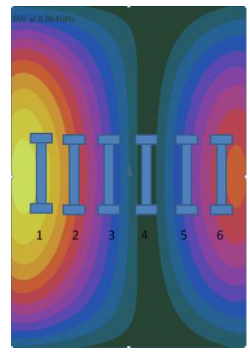

Fig. 3 Six positioning of medical stent inside ASTM phantom, the background is the electric field contour.

The incident electric field is extracted in the corresponding location. For each configuration, a heating scalar factor is calibrated using the ratio of temperature rise to the intensity of incident electric field. The heating factor for estimation employs the first configuration to minimize experimental error. The intensity of incident field is then normalized by this scalar factor to get an estimated temperature rise. A comparison of temperature rise between direct measurement and estimation by scaling method is plotted in Fig. 4. Result shows the estimation reaches a good agreement with the measurement.

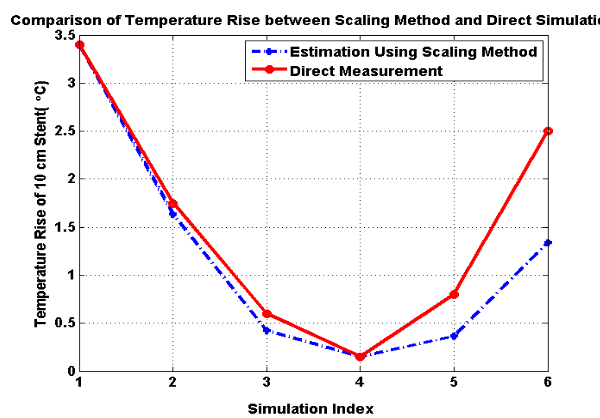

Fig. 4 Comparison of temperature rise for stent between scaling method and direct measurement [8].

C. In-vivo heating diagram

This approach is then used to estimate the temperature rises near the stent tips when the device is implanted inside a human body. This particular stent is implanted inside esophagus, aorta and intestine. Here the intestine is used as an example to demonstrate the heating.

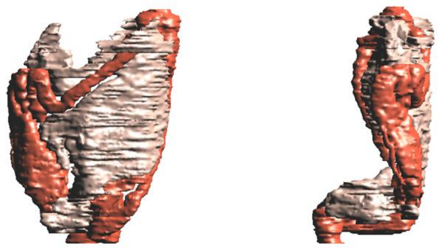

Fig. 5 The above two pictures shows the front view and side view of human intestine CAD model.

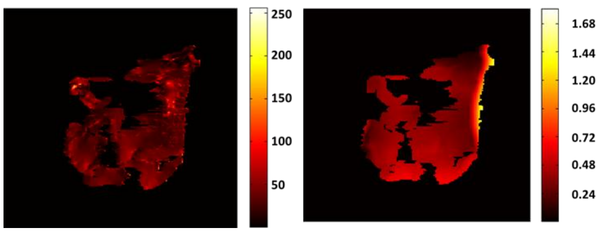

Fig. 6 The left below is the magnitude of the incident electric field on one single slide. The right below is the estimated temperature rise on the same slide.

Fig. 5 shows a front and side view of intestine. The incident electric field and temperature rise is plotted in Fig. 5. The unit of the electric field is V/m and the unit of temperature rise is °C. A highest temperature rise of 1.65 °C is obtained around boundary. This is reasonable because the field sees a discontinuity and abruptly changes across the tissue boundary.

This method suggests a way of saving us from the repeated workload of simulating all possible placement of the stent inside human body. However, this method has two limitations. Firstly the medical implants have to be electrically small compared to wavelength in order to satisfy the base assumption. Secondly, the accuracy of this may suffer from inhomogeneity of environment as human body. Further investigation is necessary to extend this method to general scenarios.

Conclusion

In the work, a scaling approach is proposed to estimate the heating of small implants. This method is validated by both numerical simulations and direct experiments of a cylindrical rod and stent inside ASTM phantom. The in-vivo RF heating performance of medical implanted stent is also investigated inside anatomic accurate human model by virtual family.

References

[1] Jin J, Electromagnetic Analysis and Design in Magnetic Resonance Imaging. Boca Raton, FL: CRC Press, 1998.

[2] T. Sommer, C. Vahlhaus, G. Lauck, A. V. Smekal, M. Reinke, U. Hofer, W. Block, F. Traber, C. Schneider, J. Gieseke, W. Jung and Schild H, ―MR imaging and cardiac pacemakers: in-vitro evaluation and in-vivo studies in 51 patients at 0.5 T,‖ Radiology, Vol. 215, No. 3, pp. 869-79, 2000.

[3] W. Kainz, G. Neubauer, R. Uberbacher, F. Alesch, and D. D. Chan, “Temperature measurement on neurological pulse generators during MR scans,” Biomedical Engineering Online, Vol. 1, No. 2, pp. 1-8, 2002.

[4] Sung-Min Park, Rungkiet Kamondetdacha, John A. Nyenhuis, “Calculation of MRI-induced heating of an implanted medical lead wire with an electric field transfer function,” Journal of Magn. Reson. Imaging. Volume 26, Issue 5, pp. 1278–1285, Nov. 2007.

[5] Liu, Y, Chen, J, Shellock, F G, Kainz, W, “Computational and experimental studies of an orthopedic implant: MRI-related heating at 1.5-T/64-MHz and 3-T/128-MHz,” J. Magn. Reson. Imaging. doi: 10.1002/jmri.23764.

[6] I Büchler P, Simon A, Burger J, Safety of Active Implantable Devices During MRI Examinations: A Finite Element Analysis of an Implantable Pump. IEEE transactions on biomedical engineering. Vol. 54, (4). pp. 726-733, Apr. 2007.

[7] SEMCAD X Reference Manual, SEMCAD Simulation Platform for Electromagnetic Compatibility.

[8] ASTM International. Standard test method for measurement of radio frequency induced heating near passive implants during magnetic resonance imaging. ASTM standard F2182-11 West Conshohocken, PA: ASTM International: 2011.