Evaluation of RF Heating of a Multi-Mode Intravascular MRI Coil

P. Wang[1], K. Kurpad[1], and O. Unal[1]

[1]Medical Physics, University of Wisconsin - Madison, Madison, WI, United States

Introduction

Accurate and reliable guidance is crucial for catheter-based treatment of cardiac diseases such as atrial fibrillation. Magnetic resonance imaging (MRI) is a preferable choice because of good soft tissue contrast, 3D capability, and lack of ionizing radiation. Higher resolution imaging of the region of interest such as arteries and pulmonary vein is desirable. Intravascular MRI coils have been proposed to achieve better SNR and increased spatial resolution [1, 2]. The goal of our research group is to develop a 3D real-time interventional MRI guidance platform with unique imaging and device tip-tracking capabilities. Studying and characterizing local heating due to eddy current from radiofrequency (RF) magnetic field is an important safety issue. Previous studies have characterized the heating from the specific absorption rate (SAR) of long wires [3]. The goal of this study is to investigate and develop a model to predict temperature rise in the vicinity of a multimode intravascular coil due to SAR as well as thermal conduction from the heated wire itself.

Materials and Methods

The schematic and circuit model of the multimode intravascular MRI coil used for the simulation studies is shown in Fig. 1 [4, 5]. It is a multimode coil consisting of a tightly wound solenoid for tip-tracking and a rectangular loop coil for imaging. The coil was modeled with 36 gauge copper wire and an acrylic tube as the core. The solenoid has 10½ turns and the rectangular coil has three turns. The entire coil was covered by polyester heat shrink for electric insulation. Circuit resonance at 64 MHz was obtained with four miniature capacitors of 47 pF. A terminal impedance Z_L is present to account for wire loss and impedance from feed line. Tissue mimicking medium was set as ambient medium around the coil. SAR, with the unit of W/kg, characterizes the power deposition in the tissue caused by the radiofrequency (RF) electric field. The wire itself can be heated by the electrical current due to non-ideal conductor and the heat can spread to the surrounding tissue by thermal conduction. Both mechanisms contribute to the temperature rise in the tissue. Induced current and generated SAR values were computed from numerical simulations with finite element method (FEM) by commercially available software (COMSOL Multiphysics v3.5a; Burlington, MA). For small coil loop with negligible phase difference in the external magnetic field, the strength of induced current can also be formulated by:

$$I = -\frac{j\omega N\Phi_1}{Z_L + Z_{coil}} \quad (1)$$

where ω is angular frequency, Φ_1 is the external RF magnetic flux through the area spanned by the loop, N is the number of wire turns. Z_{coil} is input impedance of the coil. The electromagnetic (EM) simulations were accomplished through the RF model of the FEM software with ideal conductors. Variable load impedance can be used to compensate the actual resistance of the wire, the characteristic impedance of the feed line and impedance matching network. Generated SAR can be derived from the EM solution and with coupled heat transfer model the temperature rise in time can be solved. With additional resistive heating in the wire, its contribution to the temperature rise in the tissue can also be estimated.

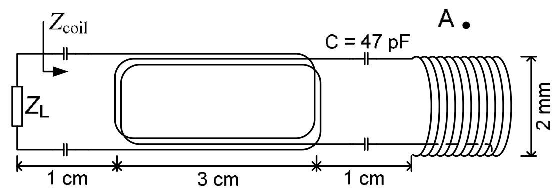

Fig. 1. Schematic of a multi-mode intravascular coil.

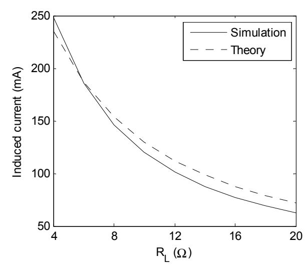

Fig. 2. Strength of induced current.

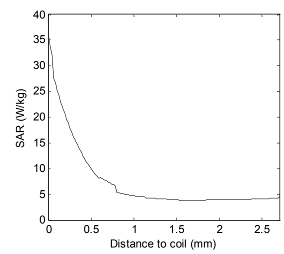

Fig. 3. SAR distribution near the tip-tracking solenoid.

Results and Discussion

In simulations, the strength of external RF field was selected such that the SAR without the multimode coil in place was about 5 W/kg. The plane of the rectangular imaging coil was placed perpendicular to the magnetic field for maximum current induction. The strength of induced current is plotted in Fig. 2 for variable resistive Z_L from 4 to 20 Ω. The theoretical curve based on equation (1) is also shown for validation purposes. For the worst case scenario, a current of 250 mA was used for the study of increased SAR and temperature. The generated SAR distribution above the tip-tracking solenoid (point A) is shown in Fig. 3.The temperature rise shown in Fig. 4 was monitored at a point 2 mm away from the solenoid (near point A). The source of resistive heating in the simulation model was derived from the strength of current and resistivity of copper wire. As seen From Fig. 4, the temperature rise caused by the generated SAR from the coil is insignificant compared to that caused by the resistive heating of the wire.

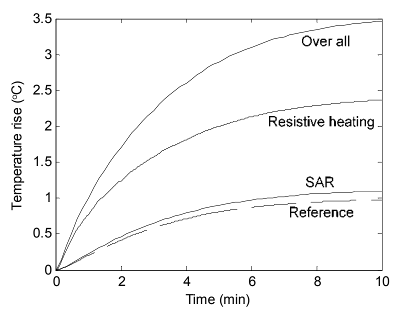

Fig. 4. Tissue temperature rise near the coil. Reference: Temperature caused by the SAR of scanner without coil; SAR: Temperature caused by total SAR of coil and scanner; Resistive heating: Temperature caused by resistive heating of copper wire alone; Over all: Over all temperature rise caused by all mechanisms.

Conclusion

Based on the results shown above, the generated SAR from this multimode intravascular MRI coil can be considered safe as long as the induced current is properly limited by coil loop impedance. The resistive heating of the wire is the leading cause for the temperature rise in the tissue. In order to limit the resistive heating, highly conductive wires are desirable for coil construction.

References

[1] Hurst G, et al, Magn Reson Med 1992; 24: 343-57.
[2] Rivas P, et al, J Cardiovasc Magn Reson 2002; 4: 223-32.
[3] Yeung C, et al, Magn Reson Med 2007; 58: 1028-34.
[4] Unal O, et al. ISMRM **1398**, 2006.
[5] Kurpad K, et al. ISMRM **292**, 2007.