Electromagnetic and Thermal Modeling of SAR and Temperature Fields in Tissue Due to an RF Decoupling Coil

Jeffrey W. Hand,¹ Raymond W. Lau,² Jan J.W. Lagendijk,³ Junxiao Ling,⁴ Mike Burl,⁴ and Ian R. Young⁵

The finite difference time domain method is used to calculate the specific absorption rate (SAR) due to a butterfly surface coil in a realistic tissue model of the leg. The resulting temperature distribution and temperature changes are found using a finite difference solution to the bioheat transfer equation. Reasonable agreement is found between predicted temperature changes and those measured in vivo provided that the resulting hyperthermia does not induce noticeable changes in perfusion. The method is applicable to radiofrequency dosimetry problems associated with high B₀ field magnetic resonance systems and where knowledge of spatial variation in SAR is important in assessing the safety of new magnetic resonance procedures.

Key words: FDTD; RF absorbed power; MR safety; thermal model

To avoid potential thermal hazards related to exposure to high-intensity radiofrequency (RF) fields that might occur during MR procedures, national agencies have issued guidelines describing recommended limits. These limits are expressed in terms of increase in body temperature and restrictions on the maximum temperatures within local regions of the body (e.g., head, trunk, limbs) (1–4). Since spatial variations in internal body temperature are difficult to measure, and the power absorbed in tissue is at least in principle easier to calculate, these guidelines recommend secondary restrictions in terms of specific absorption rate (SAR). For example, the limits to the SAR averaged over the whole body recommended by the NRPB (2) are 1 W/kg for periods in excess of 30 min, 2 W/kg for periods shorter than 15 min, and 30 W min/kg for periods of intermediate duration. Restrictions on local SAR (averaged over 1 kg of tissue and over any 6 min period) are 2, 4, and 6 W/kg for head, trunk, and limbs, respectively. These limits are doubled if the total period of exposure is less than 30 min. The US Food and Drug Administration (FDA) (1) considers a “significant risk” to be present in MR investigations in which the SAR is greater than 4 W/kg whole body for 15 min, 3 W/kg averaged over the head for 10 min, 8 W/kg in any 1 g of tissue in the head or torso for 15 min, or 12 W/kg in any 1 g of tissue in the extremities for 15 min.

There have been several reports describing temperature changes induced in patients and volunteers undergoing MR procedures. For example, Shellock et al. (5) measured core and skin temperatures in healthy volunteers exposed to whole body averaged SARs of 2.7–4 W/kg during a 30 min procedure in a 1.5 T scanner. The highest skin temperatures and statistically significant temperature increases recorded on the upper arm, forearm and chest were 38.1°C, 36.0°C, and 34.5°C and 7.5°C, 3.9°C, and 2.9°C, respectively. The greatest increase observed in core temperature was 0.2°C. In another investigation (6), core and skin temperatures were monitored in 50 patients undergoing imaging of the trunk in which whole body averaged SARs ranged from 0.42 to 1.2 W/kg. A statistically significant increase in core temperature was observed following imaging (mean increase 0.2°C, range 0.0–0.5°C), the greatest increase in skin temperature was 3.5°C, and the highest skin temperature was 35.1°C. In a later study (7) Shellock et al. studied volunteers undergoing imaging performed at whole body averaged SARs of 6.0 W/kg in cool and warm environments. Although heating of surface tissues appears to be associated with MR procedures, there is a paucity of data regarding temperature changes in deeper tissues in human subjects.

Consequences of a lack of thermoregulation during exposure to RF fields from 1.5 T imagers were highlighted by studies carried out on anesthetized animals. For example, in dogs, SARs in the range 5.9–10.5 W/kg led to site-specific (including deep tissues) mean temperature increases of up to 4.6°C (8). In sheep, whole body SARs in the range 1.5–4 W/kg led to temperature increases in subcutaneous and deep tissues, although these did not produce adverse thermal effects. However, it was postulated that such effects might be produced at SARs of around 8 W/kg (9).

Early studies describing fields and power absorption associated with MR RF coils were often based on geometrically simple homogeneous models (10) or inhomogeneous phantoms (11). While such studies provide valuable insight and reasonable estimates of spatially averaged power absorption during MR procedures, they cannot reliably predict local peak SAR levels in the dielectrically heterogeneous human body (12). Grandolfo et al. (13) used a quasi-static method to model power absorption from an idealized 63 MHz magnetic field in which the phase of the field was constant across the body. However, the assumption that the RF field is not perturbed by the body, inherent in the quasi-static approach, leads to an overestimation of
SAR at higher frequencies (14). More recently, analyses of fields produced by saddle coils operating at 64 MHz in models of the pelvis and abdomen (15) and head (16) have been reported. These studies used a modified formulation of the quasi-static calculation of eddy currents in which a complex conductivity term accounted for both conduction and displacement currents. Recently Jin et al. (17) used a biconjugate-fast Fourier transform (BCG-FFT) method, and Chen et al. (18) used a combination of the method of moments and the finite difference time domain (FDTD) method to compute the electromagnetic fields of RF coils loaded with an anatomically accurate model of the human head. Calculations were made for frequencies between 64 and 256 MHz.

In general, accurate modeling of SAR in the human body associated with RF fields with frequencies greater than 40–60 MHz requires a full, time-dependent solution of Maxwell’s equations since the wavelength in tissue is then significant compared with body dimensions and conduction and displacement currents lead to changes in the RF field within the subject. Several numerical methods for solving Maxwell’s equations and calculating the spatial dependence of SAR in the body or body regions, involving both frequency and time domains, have been investigated. Of these the FDTD method (19,20) offers several advantages for modeling RF propagation and has been used to solve a range of biomedical problems (21,22), including the investigation of RF coils for MR applications (23–25).

Thermal sequelae of RF power deposition associated with MR procedures have been studied using compartmental models that describe human thermoregulation (26). However, with this approach the effects are based on SAR averaged over the volume of the compartment and so details of local temperature fields cannot be predicted. Models that describe the detailed local heat transfer within tissues are more suited to predicting local temperature fields and simulations of this type are usually based on the approach suggested by Pennes (27) in which the effects of perfusion on the temperature distribution are accounted for by a heat sink term (28).

In this work we use the FDTD method to calculate the electromagnetic field associated with a 64 MHz butterfly surface coil designed for decoupling during 13C spectroscopy and the associated spatial variation in SAR induced in a dielectrically heterogeneous model of the human leg. Resultant temperature changes in the leg are predicted using a finite difference solution for heat transport within tissue. Predicted temperature changes are compared with measurements obtained while the leg of a volunteer was exposed to the RF field from the coil.

**MATERIALS AND METHODS**

**Electromagnetic Modeling**

The FDTD technique has been described in detail elsewhere (19,20). For the sake of completeness, only a brief outline of the technique is given here. The pair of Maxwell’s equations

\[ \nabla \times \vec{E} = -\mu \frac{\partial \vec{H}}{\partial t} \]  

(1)

\[ \nabla \times \vec{H} = \epsilon \frac{\partial \vec{E}}{\partial t} + \sigma \vec{E} \]  

(2)

(where \( \vec{E} \) and \( \vec{H} \) are vector quantities) is approximated by an explicit finite-difference scheme in 3-dimensional (3D) rectangular Cartesian coordinates. The coil and body part under study are represented on the grid by a collection of cells that are segmented according to the relative permittivity \( \epsilon \) and electrical conductivity \( \sigma \) of the different tissue media. The E- and H-field components are calculated with respect to each cell in the manner described by Yee (19). Following switch-on of the field source(s), the propagation and scattering of the field throughout the grid is modeled by alternately time-stepping the finite difference expressions for all E- and H-field components until steady-state conditions are achieved.

In this work, steady state was deemed to have been achieved when the computed fields differed by no more than 0.3% when compared with the values obtained in the preceding cycle. Second-order radiation boundary conditions (29) were enforced at the edges of the grid to absorb the outgoing field with minimal reflection there. The resulting local E-field at any cell location was found from the square root of the sum of the squares of the \( E_x, E_y, \) and \( E_z \) components determined for that cell. The local SAR was found using Eq. [3]:

\[ \text{SAR} = \frac{\sigma |\vec{E}|^2}{2\rho} \ \text{W/kg} \]  

(3)

where \( |\vec{E}| \) is the magnitude of the local E-field and \( \rho \) and \( \sigma \) are the local tissue density and electrical conductivity, respectively.

The work described here was carried out using an FDTD code written in FORTRAN 77 (30). The code has been developed to meet several applications and has been verified in a number of ways over more than a decade. For example, in early testing and verification the code was used to calculate fields and SAR due to a 900 MHz dipole, 0.4 × wavelength long, and placed 15 mm from spheres and cubes simulating the head. Results (E-field magnitude, SAR) using these standardized models were compared with other FDTD codes and other computational methods as part of the European Cooperation in the field of Scientific and Technical Research (COST) programme (31,32). Verification of the code for applications in clinical hyperthermia has also been made (33) and recently we have further verified the use of this code by comparing numerically predicted E-fields due to a rectangular coil in a phantom experiment with those measured using a calibrated minimally perturbing E-field probe in an identical experiment (24,34).

The butterfly coil for proton decoupling studied in this work is shown schematically in Fig. 1. In the FDTD model, a single plane of cells represented the copper conductors. These cells had \( \epsilon = 1 \) and \( \sigma = 10^2 \) S/m and the metallic boundary conditions imposed were that the E-field component normal to the plane of the coil was reflected while that in the plane of the coil was set to zero. The gaps in the coil
structure and other air regions were represented by cells for which \( \varepsilon' = 1 \) and \( \sigma = 0 \) S/m. The sources in the model were taken to be sinusoidally time-varying 64 MHz E-fields excited across the 2 cm gaps in the coil structure. The amplitudes of these were set to 100 V/m in the model; higher powers were simulated subsequently by applying a scaling factor.

A model of the left leg of a volunteer was developed from a segmented MRI data file consisting of 79 planar sections each of 256 \( \times \) 256 pixels. The pixel dimensions were 1 \( \times \) 1 mm, the thickness of the sections was 5 mm, and the pixels were segmented to air, subcutaneous fat, internal fat, bone, and muscle. A new data set with dimensions 51 \( \times \) 51 \( \times \) 79 was generated for the FDTD model by decreasing the in-plane resolution to 5 mm (by averaging within the plane) and segmenting to muscle, fat/bone and air. The dielectric properties allocated to these 5 \( \times \) 5 \( \times \) 5 mm\(^3\) voxels are shown in Table 1. Tissues with low water content such as bone, fat, and internal fat were not distinguished. No averaging was performed between sections and since averaging within the sections did not lead to a wide spread of dielectric properties, no intermediate dielectric properties were required.

The leg and butterfly coil were positioned relative to each other and to the boundaries of the computational space as shown in Fig. 2. At their nearest approach, the coil and leg were separated by one cell. The model was run with 1874 time steps per cycle (\( \Delta t = 8.34 \times 10^{-12} \) sec) to satisfy stability criteria. Steady-state conditions were reached after 10 cycles. Computations were performed using a Silicon Graphics Indigo workstation (R4400 150 MHz CPU, 128 MB RAM).

Thermal Modeling

Temperature distributions in the leg under normal conditions and due to energy absorption from the RF coil were calculated using a finite difference numerical model (35) based on the form of bioheat transfer equation (Eq. [4]) suggested by Pennes (27), namely

\[
\rho c \frac{dT}{dt} = \nabla \cdot (k \nabla T) + P + B
\]

where \( \rho \), c, and k are the density, specific heat, and thermal conductivity of the tissue, \( T \) is the local temperature, \( t \) is the time, and \( P \) represents the power per unit volume absorbed from the coil. Metabolic heat production was neglected since the values for fat and muscle (350, 700 W/m\(^3\), respectively) are small compared with \( P \). The term B represents the rate of loss of energy per unit volume due to

<table>
<thead>
<tr>
<th>Medium (as segmented in original MR data set)</th>
<th>FDTD model type</th>
<th>Relative permittivity ( \varepsilon' )</th>
<th>Electrical conductivity ( \sigma ) (S/m)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Air</td>
<td>1</td>
<td>1.0</td>
<td>0</td>
</tr>
<tr>
<td>Bone</td>
<td>3</td>
<td>8.5</td>
<td>0.06</td>
</tr>
<tr>
<td>Muscle</td>
<td>9</td>
<td>72.0</td>
<td>0.8</td>
</tr>
<tr>
<td>Fat</td>
<td>3</td>
<td>8.5</td>
<td>0.06</td>
</tr>
<tr>
<td>Internal fat</td>
<td>3</td>
<td>8.5</td>
<td>0.06</td>
</tr>
<tr>
<td>Copper conductor</td>
<td>2</td>
<td>1.0</td>
<td>( 10^5 )</td>
</tr>
</tbody>
</table>

FIG. 1. Schematic diagram of 64 MHz butterfly coil. All measurements are in cm. The gaps in the coil structure are 2 cm wide and 67 pF capacitors are connected across them. The arrows indicate the relative directions of the electric fields across the gaps.

FIG. 2. Orthogonal views of the leg and coil within the boundaries of the computational space of 51 \( \times \) 66 \( \times \) 89 cells, each 5 \( \times \) 5 \( \times \) 5 mm\(^3\). The minimum separation between the leg and coil is 1 cell. The positions of sections A, F, and K referred to in Figs. 4–8 are indicated.
blood flow and this was taken to be

\[ B = -w_{\text{blood}}(T - T_{\text{core}}) \tag{5} \]

where \( w \) is the volumetric perfusion rate, \( c_{\text{blood}} \) is the specific heat of blood, and \( T_{\text{core}} \) represents the core temperature in the leg. The value of \( w \) was assumed to be constant and independent of temperature in view of the relatively small changes in temperature generally expected in this study. The values of the thermal parameters used in these calculations were based on those reported in (36) and are given in Table 2.

Tissue segmentation and the spatial resolution of 5 mm matched exactly those used for the FDTD electromagnetic model. The heat transfer coefficient from the outer surface of the tissues to the surrounding air was taken to be 20 W/m²/°C. Modeling the effects of discrete vessels was not taken into account. Temperature distributions were calculated at 60, 300, 600, 900, 1200, 1800, and 2400 sec after power on as well as for the steady-state case (3600 sec). The distribution of temperature changes due to energy absorption from the RF field was calculated from the difference of the temperature distributions calculated for the cases of power on and power off.

Experimental Measurements

Experimental verification of some of the predictions made using the electromagnetic and thermal models described above was also undertaken. Experiments were performed in which optical fiber temperature sensors (Luxtron 3000 system, Luxtron, Mountain View, CA) were placed on the leg of a volunteer's leg (the same leg that had been imaged previously) at the location of the section in which the maximum temperature was predicted and 10 cm from that position both toward the thigh and the ankle. Temperatures were also measured on the heated leg on the side opposite to the coil and on both outer and inner sides of the unheated leg as controls. The butterfly decoupling coil was placed adjacent to the leg in such a way as to simulate the conditions assumed in the model.

A schematic representation of the experiment is shown in Fig. 3. The coil was tuned and matched to provide a return loss of less than -20 dB at 64 MHz and was driven in a pulsed mode at a 10% duty cycle using a frequency synthesizer (Hewlett Packard model HP3335A), a laboratory signal generator, and a 5 kW linear amplifier (ENI model 5000PK). The signal generator output was adjusted such that the power delivered to the coil was 100 W (10 W time averaged). Measurements of the coil losses showed a change of 5:1 between loaded and unloaded conditions, suggesting that approximately 8 of the 10 W delivered to the coil were absorbed in the leg. This estimate was in agreement with measurement of the field produced by the coil using a flux loop 12 mm in diameter. From this measurement and the calculated sensitivity of the coil at the location of the flux loop beneath the center of the coil, the current through the coil and hence the voltage across each of the 67 pF capacitors placed across the gaps in the real coil could be estimated and, allowing for duty cycle, compared with the driving voltages assumed in the simulation.

### RESULTS

#### Predicted SAR Distribution

Figure 4 shows the distribution of high and low water content (grayscale images) in 11 cross sections of the leg. The separation between these sections is 2 cm. The proximity of the decoupling coil to the leg is also indicated. Also shown is the distribution of SAR in each section, normalized to the global maximum calculated in the leg that occurred in section F. The sections shown cover almost the entire region in which a SAR of at least 50% of the global maximum is present (the boundaries of this region are formed by section A and 5 mm beyond section K). The asymmetry of the SAR distribution with respect to the coil occurs because the surface of the leg and the coil are not parallel. It can also be seen from Fig. 4 that the SAR distribution is strongly dependent on a) the spatial distribution of muscle and fat/bone and b) the separation between the coil and the leg. In general, since currents in the tissue tend to pass through the more conducting muscle, it is in this tissue that the higher SARs occur. It can also be seen that some coupling of the fields occurs at the distal side of the leg, giving rise to an SAR in the region of 10% of the peak value.

#### Predicted Temperature Distributions

Figure 5 shows the predicted steady-state temperature distribution in the absence of RF power predicted in each of the sections A–K referred to in Fig. 4. These results are based on the assumptions that the temperature of the environment was 20°C, the heat transfer coefficient between tissues and the environment was 20 W/m²/°C, and the “core” temperature (used in the heat sink term of the bioheat transfer equation) was 37°C.

Figure 6 shows the predicted steady-state temperature distribution in the same sections in the case in which 2 W of RF power were absorbed in the leg. The total power absorbed was found from a summation of the SAR values in each of the 21,158 5 × 5 × 5 mm cells within the leg model. The global maximum SAR within a cell for this total absorbed power was 12.8 W/kg. Since some safety guidelines refer to SAR averaged over 1 g of tissue (1,4), the SAR spatially averaged over 8 cells (1 cm³, approximately 1 g of tissue) was calculated. The maximum value was 9.1 W/kg. Although an increase in temperature was observed, the maximum temperature remained at 37°C and this occurred within the central region of the sections. In the case of 8 W
absorption (not shown), the predicted maximum temperature was 39.4°C and, like the global maximum SAR (51.2 W/kg), occurred in muscle at the muscle/fat interface in cross section F. The maximum SAR averaged over 1 cm³ was 36.6 W/kg and also occurred in this cross section.

Figure 7 shows the temperature changes predicted for the 2 W exposure conditions (difference between steady-state temperatures with and in the absence of this level RF power). The predicted maximum change in temperature is 1.3°C and occurs in the same section as the maximum SAR.

FIG. 3. Schematic diagram of experimental set-up for exposing a volunteer’s leg to the RF field of the 64 MHz butterfly coil.

FIG. 4. Distribution of fat/bone (light shading) and muscle (dark shading) (upper images) and corresponding SAR distributions (lower images) in 11 sections of the leg. The separation between sections is 2 cm. The SAR is normalized to the global maximum, which occurs in section F. Areas in which the SAR < 0.1% of the maximum SAR appear as the background color in these images. The relative positions of coil and leg are indicated in the upper images.
Temperature changes predicted for the 8 W are 4 times those shown in Fig. 7.

Measured Temperature Changes In Vivo

Figure 8 shows a comparison of predictions and temperature changes recorded on the skin of a volunteer’s leg heated by the decoupling coil in the experimental simulations of the model. Two experiments were carried out, at 2 and 8 W absorbed power, respectively. The experiment in which the leg was subjected to 8 W absorbed power was terminated before steady state was reached because of the volunteer’s discomfort. The locations “central,” “toward thigh,” and “toward ankle” were representative of sections F, A, and K, respectively, shown in Figs. 4–7. An attempt to correct for changes in temperature due to causes other than RF heating was made by observing the changes in skin temperature on the unheated right leg and subtracting these from the overall changes observed in the heated leg. Predicted temperature changes shown are the maximum and that calculated at the center of a voxel located on the superficial fat layer in cross section F and the maxima (8 W experiment) and at the center of voxels located in the superficial fat layer (2 W experiment) in sections A and K.

The locations at which these predictions were made are shown on the three sections in Fig. 9.

DISCUSSION

We have used the finite difference time domain method to compute the RF electromagnetic fields from a 64 MHz decoupling coil leading to the prediction of SAR within a model of the human leg placed in proximity to the coil. The predicted SAR distribution is strongly dependent on the spatial distribution of muscle and fat/bone within the leg and the separation between the coil and the leg. The lowest SAR in a leg cross section does not necessarily occur at the region of the leg cross section most distal from the coil.

It has been shown previously (24) that predictions of fields associated with an RF coil made using the FDTD code are in good agreement (within 1 dB) with measured fields within a phantom. A further approximation has been made in the present work in that we have assumed that the values of the dielectric properties of the tissues are those given in the literature (13). However, the variations in reported dielectric properties are unlikely to change the qualitative nature of the predicted SAR distribution, and as such, the method should prove useful in assessment of RF safety for MR applications. A parametric study of the sensitivity of predicted temperature changes resulting from...
exposure to electromagnetic fields from hyperthermia devices (70 MHz) suggests that an uncertainty of 20% results by varying dielectric properties by a factor of two (Lavies (70 MHz) suggests that an uncertainty of 20% results by varying dielectric properties by a factor of two (Lavies (70 MHz) suggests that an uncertainty of 20% results by varying dielectric properties by a factor of two (Lavies (70 MHz) suggests that an uncertainty of 20% results by varying dielectric properties by a factor of two (Lavies (70 MHz) suggests that an uncertainty of 20% results by varying dielectric properties by a factor of two). Morvan et al. (37), using an MRI diffusion technique, reported similar temperature changes induced by a butterfly coil approximately 23 × 29 cm in size and driven at 86 MHz. Experiments in which 6 W input power was applied to the coil resulted in a temperature change of 4.9 ± 1.9°C in an area of calf muscle close to the skin in a volunteer. Rather than use MR-based temperature sensing, which may be subject to several sources of artifact and uncertainty (38,39), we chose to use direct temperature measurement using fiberoptic temperature sensors to obtain experimental data for comparison with predictions. Direct measurement has the considerable advantage of unequivocal accuracy at spot locations but the disadvantage of only providing sparse spatial sampling.

Transient temperature predictions were in reasonable quantitative agreement with measurements when the temperature change was such as to be unlikely to cause a large change in perfusion. For example, the results of the 2 W experiment in Fig. 8 show that predictions and measurements exhibit similar temperatures and initial rates of change of temperature (and hence local SARs). There was a tendency for the predictions to underestimate measurements by approximately 0.2°–0.3°C at times beyond about 30 min into the experiment. A small continuous increase in skin temperature was observed in the experiments at times as long as 50 min. This could have been due to the difficulty in stabilizing the leg temperature to better than 0.2°C over such periods, rather than being a local consequence of the SAR due to the coil. During the 8 W experiment, the relative differences between the measurements made at sections A and K and that at section F were smaller than those seen in the 2 W experiment and the results were qualitatively different, suggesting a change in the heat transfer mechanism. Although the predicted transient temperature at the fat layer in cross section F was comparable to that observed on the skin in that section, the observed changes in cross sections A and K were greater than predicted changes, including the predicted maximum changes within these sections. It was observed that the skin on the volunteer's leg became hyperemic during the 20 min exposure to 8 W of RF. Under these conditions, the present thermal model, which assumes constant perfusion and isotropic heat transfer, appears to provide an inadequate approximation to conditions occurring in the leg under the conditions of thermally triggered perfusion changes (40). Another difficulty in making a quantitative comparison between predictions and measurements in these experiments is that the measurements were made on the volunteer's skin while the predictions are at the centers of voxels in the superficial fat layer.

FIG. 7. Predicted steady-state temperature changes in sections A through K due to 2 W power absorption in the leg. The color scale covers the range 0°–1.3°C. The maximum change is 1.3°C and occurs at the muscle/fat interface in cross section F.

The quest for accurate quantitative agreement between predicted temperature changes and real ones is challenging in view of the number of parameters that influence the predicted values. In particular, quantitative knowledge of the perfusion as a function of both tissue type and time as well as the heat transfer coefficient between the tissues and the environment is important. Non-isotropic heat transfer may also need to be taken into account. In the present work, which is a preliminary study using a thermal modeling approach, we simply assumed basal values of perfusion reported in the literature together with a realistic value of the heat transfer coefficient h for bare, dry skin in the absence of strong draughts. Nevertheless, there is reasonable agreement between the experimentally observed and theoretically predicted temperature changes in the absence of noticeable hyperthermally induced changes in perfusion. Despite the relatively large maximum temperature change of 5.2°C predicted in the leg muscle due to 8 W absorbed power at 64 MHz, the maximum predicted temperature (39.4°C) remains within NRPB guidelines (2) since the basal temperature of the muscle in this region was predicted to be 33.2°C under the assumed conditions of perfusion and heat transfer at the tissue surface.
The available RAM in the computing resource used in this work (Silicon Graphics Indigo, R4400 150 MHz CPU, 128 MB RAM) imposed the practical upper limit on the number of cells within the FDTD model. A spatial resolution of 5 mm was chosen for the present leg model to be compliant with this upper limit yet comparable to those reported in other studies (18,25). The FDTD method is computationally efficient in terms of computer memory requirement and ease of grid construction and a spatial resolution of 1–2 mm in SAR predictions is achievable over appropriate tissue volumes with only moderate improvement over the workstation used in the work reported here. Although some improvement in the spatial resolution of SAR distributions may be desirable, for example, to account for skin and in the case of higher frequencies, it should be noted that the Pennes form of the bioheat transfer equation (27) cannot account for spatial variations in temperature that occur over distances of a few millimeters or less and that are due to the effects of discrete blood vessels. Thus, it may not be realistic to perform temperature calculations at significantly better spatial resolution.

FIG. 8. Comparison of measured and predicted temperature changes due to RF heating from 64 MHz coil. Upper graph: experiment in which power absorbed in leg was 2 W. The open symbols refer to measurements made on skin (open box, in cross section F; open triangle, toward thigh, in section A; open circle, toward ankle, in section K), and the closed symbols refer to predicted temperature changes (closed box, fat surface in cross section F; closed triangle and circle, predicted changes at fat surface in sections A and K, respectively. Lower graph: experiment in which power absorbed in the leg was 8 W. The additional symbols + and × represent predicted maxima in the sections A and K, respectively. The measured temperature changes are corrected for changes observed in the non-heated leg. The locations in the three cross sections of the predicted changes are shown in Fig. 9.

FIG. 9. Locations of predicted temperature changes in cross sections A, F, and K (central, toward thigh and toward ankle, respectively) referred to in Fig. 8. Open circle, maximum predicted change; open box, predicted change at fat surface. The locations of the predicted temperature changes at the model surface approximated the locations at which the measurements were made on the skin of the volunteer's leg.
without accounting for discrete vasculature. Finally, high spatial resolution in temperature predictions is only required when the characteristic dimensions of the heat source are equally small.

The development of the modeling tool described above is timely since the increasing availability of higher $B_1$ field systems and procedures such as proton decoupling and imaging techniques with multiple RF excited echoes may lead to safety guidelines being challenged. There is therefore an increasing need to know the spatial distribution of SAR and related temperature changes during such procedures. Similar needs arise with the advent of intra-cavitary transmit coils and related devices since the use of small transmit coils in close proximity to tissue results in extremely non-uniform SAR, which cannot be interpreted by conventional methods of assessing safety.

CONCLUSIONS

A method for predicting SAR associated with a 64 MHz decoupling coil in realistic tissue model of the human leg has been developed. Since a full time-dependent solution of Maxwell’s equations is implemented, the technique is applicable to problems associated with high $B_1$ MR systems. The temperature changes caused by RF power absorption in the tissues are predicted using a thermal model. These predictions are in reasonable agreement with measurements made in vivo provided that the resulting hyperthermia does not induce noticeable changes in perfusion within the tissue.

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REFERENCES

36. Hand JW, Ledda JL, Evans NTS. Considerations of radiofrequency


