Electromagnetic and heat transfer computations for non-ionizing radiation dosimetry

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Abstract. Reliable information on the heat distribution inside biological tissues is essential for the planning and optimization of experiments which aim to study the effects of non-ionizing radiation (NIR). In electrodynamics, the finite-difference time-domain (FDTD) technique has become the dominant technique for radiofrequency dosimetry. In order to obtain the electromagnetic field and heat distributions within the same simulation run without changing discretization, a heat diffusion solver has been directly integrated into an advanced electrodynamic FDTD kernel. The implementation enables both coupled and sequential simulations. It also includes the ability to work with complex bodies and to accelerate heat diffusion. This paper emphasizes the importance of this combination in the field of NIR dosimetry. Two examples from this area are given: the validation of dosimetry with temperature probes and the estimation of the highest thermal load during bioexperiments.

1. Introduction

The determination of the electromagnetic absorption inside biological tissues for a given excitation, usually expressed in terms of the specific absorption rate (SAR), is still a highly complicated and challenging scientific task. Among the various numerical electrodynamics techniques proposed for the calculation of SAR, the FDTD method has clearly become the most widespread due to its robustness and ability to discretize strongly inhomogeneous structures in a straightforward manner, as described in Taflove (1995). The temperature distribution as a consequence of electromagnetic power absorption is determined by the heat transfer mechanisms of conduction, convection and radiation. In living bodies other mechanisms of active heat generation or dissipation, such as metabolism, blood flow, evaporation (sweating) etc, are equally important for the resulting temperature distribution. Although the time scale of their dynamics is substantially different, the processes of electromagnetic absorption and heat transfer may become interdependent, if there is a significant temperature elevation inside the body. In this case the dielectric properties of tissues may vary with temperature leading to a different deposition pattern of the electromagnetic energy.

A relatively large body of literature on the heating up of tissues is available in the area of hyperthermia studies (Sullivan 1990, Cresson et al 1994). On the other hand, unwanted temperature increase in human tissue may occur in the close vicinity of other applications, e.g. transmitters, radar, RF sealers, magnetic resonance imaging devices, etc. For these cases,

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safety limits have been defined in terms of SAR in order to prevent health hazards caused by local or whole-body temperature increases (Kuster et al 1997). The respective temperature rise has been calculated in a number of studies concerning the electromagnetic safety of medical (Hand et al 1999) and telecommunications (Taflove and Brodwin 1975, Mokhtech et al 1994, Anderson and Joyner 1995, Lu et al 1996, Bernardi et al 1998, Wang and Fujiwara et al 1999, van Leeuwen et al 1999, Riu and Foster 1999) equipment.

In the area of bioelectromagnetics, thermal consideration is crucial in the context of dosimetry for bioexperiments. One experimental method to determine the local SAR values is the use of small temperature probes or thermographic cameras. The low sensitivity of such probes (>0.1 mK s\(^{-1}\)) often requires measurement periods of relatively long duration (from a few seconds up to several minutes), during which the SAR values may change substantially due to thermodynamic processes. Thus, it is important to be able to take into account these processes in order to validate the outcome of the experimental dosimetry with thermometry. In a recent publication Moros and Pickard (1999) have shown that the thermometrically estimated SAR is influenced by thermal conduction to a large extent. In order to restrict the errors from heat diffusion within acceptable levels, they have given a rule of thumb which the measurement time should follow during thermal SAR assessment. In another publication by the same group (Pickard et al 1999) the authors have investigated the contribution of heat diffusion in the dosimetry of a culture flask exposed to NIR. They have concluded that point SAR measurements performed with traditional thermometry methods are subject to significant errors and proposed a new experimental method for thermally inferring SAR with a differential thermometer.

Another point of interest for the planning and optimization of bioexperiments is the maximum expected thermal load which may be induced in the biological tissues. This information is important especially for experiments which are set up to test the effect of low-level NIR exposure and is often lacking in accounts of published experiments.

The objectives of our study were (a) to integrate a thermodynamic solver into an advanced electromagnetic FDTD kernel, providing easy handling and new capabilities, and (b) to demonstrate the use of such a tool in the area of NIR dosimetry.

2. Electromagnetic scattering

The thermal simulation module was incorporated in the software kernel of EMLAB (ISE AG, Balgriststrasse 102, CH-8008 Zurich, Switzerland). EMLAB is an advanced electromagnetic solver based on the FDTD method, i.e. it is a time domain code which can directly compute the evolution of the electromagnetic field. Nevertheless, it includes other modes of simulation, like static and quasi-frequency domain operations, which allow the use of multifrequency excitations and the extraction of quantities in the frequency domain.

During transient simulations all field quantities, as well as integral quantities like voltage and current, can be monitored at pre-selected locations. The excitation of the electromagnetic field can be any of the known types in FDTD, i.e. ‘hard’ source, ‘soft’ source and current density source. It can also take the physical form of a voltage source, an electric dipole, a TEM excitation or a plane wave. The temporal behaviour of the excitation signal can be chosen from various types, such as harmonic, pulse and step functions as well as modulated signals.

EMLAB supports several boundary conditions for the truncation of the computational domain, i.e. Mur’s absorbing boundary conditions, Hidgon’s operator, Berenger’s perfect matching layer (PML) and periodic boundaries. It also works with both uniform and graded meshes, allowing for an optimum utilization of the available computer resources with respect to the discretized model.
3. Heat transfer

3.1. Physical mechanisms considered

Biological bodies interact thermally with their environment with several physical and physiological mechanisms. Inside the tissues the mechanisms of conduction and convection are mainly responsible for the temperature distribution: heat conduction is present in all situations, whereas convection, in the form of blood flow, exists only inside living bodies.

The studies which examine the safety of NIR exposure have to take into account all respective heat transfer mechanisms. This is a difficult task due to the variability of biological parameters, such as the blood perfusion rate and the blood temperature, and the nonlinear nature of physiological processes. Although many models have appeared in the literature for the description of heat transfer in tissues (Arkin et al. 1994), the bioheat transfer equation (BHTE) (Pennes 1948), has been the most widely used. In some studies on the exposure to NIR (Taflove and Brodwin 1975, Bernardi et al. 1998), the convective term of the BHTE is left out, so that the latter reduces to a heat diffusion equation.

Contrary to the studies on the safety of NIR exposure, in our study which refers to dosimetry it is not necessary to consider all the heat transfer mechanisms involved. The first application which is of interest to our work is the investigation of the effect of heat transfer on the dosimetry performed with temperature probes. Both examples which will be presented concern in vitro dosimetry, therefore only heat conduction has to be considered.

The estimation of the maximum thermal load that will be delivered to the biomaterials (e.g. cell solutions, animal tissues, etc) is another application of the developed software in the context of low-level exposure bioexperiments. In this case, the calculation of the steady-state temperature distribution inside biological bodies in vivo would require the use of a model like the BHTE, according to the above discussion. Nevertheless, in both in vitro and in vivo bioexperiments it is equally important to be able to predict a priori the highest temperature rise that will appear in the biomaterials in order to facilitate the determination of the exposure levels that can elicit thermal effects. The worst-case approach which aims at a conservative estimation of the maximum permissible exposure level for studying non-thermal effects of NIR can be based on the exclusion of heat exchange mechanisms like convection and radiation. Hence, blood flow inside the biological bodies can be omitted and insulation at their boundaries can be assumed.

Insulating boundary conditions are also used in the first application which examines the effect of heat conduction on SAR assessment with thermometry. If convective or radiative boundary conditions were used instead, then the results, especially close to the boundaries, would contain errors which would reflect the contribution from all heat transfer mechanisms and not just heat diffusion, as it is intended.

3.2. Implemented solution method

The temperature rise inside the biological materials is calculated in our study using the heat conduction equation

$$\rho c \frac{\partial T}{\partial t} = \nabla (k \nabla T) + Q_{EM}$$

where $T$ is the temperature, $\rho$ is the mass density, $c$ is the specific heat, $k$ is the thermal conductivity and $Q_{EM}$ is the heat generation rate due to the deposited electromagnetic power. Equation (1) can be solved numerically with various techniques, which differ in the way the discretization equations are derived. In the developed thermal solver an explicit
finite-differencing scheme was implemented, which was derived with the ‘control-volume formulation’ (Patankar 1980). This approach has been chosen because it is compatible with the FDTD discretization. Moreover, it enables a simple treatment of boundary conditions. Each cell of the electromagnetic FDTD solving scheme is considered as a ‘control volume’ with homogeneous physical properties. The temperature node is at the centre of the control volume and linear variation of temperature between nodes is assumed.

Biological media are highly inhomogeneous, so thermal conductivity varies for every control volume in the general case. Hence, to assure self-consistency of the algorithm for the heat flux at the surfaces between neighbouring cells, the weighted thermal conductivity at the faces of the control volume is used in the discretized equations.

On the outer surfaces of the objects various boundary conditions can be applied by assuming them to have nodes, which correspond to cells of zero volume. All types of boundary conditions have been implemented in the software apart from the radiative boundary condition. The latter has not been treated separately, because under certain conditions it can be transformed to a convective boundary by introducing an equivalent radiative heat transfer coefficient as proposed in Özişik (1993).

Numerical stability of the algorithm can be guaranteed by applying the second law of thermodynamics, which requires heat to flow downhill on the temperature scale. It is not necessary to undertake a von Neumann stability analysis in this case, since physical implausibility is detectable. Thus, the stable time step is found before the simulation starts by examining every cell of the discretized space and enforcing the multiplication coefficient of the temperature term at the current time step to be positive; otherwise the value of temperature at the next time step would become lower than the temperature at the current time step, despite the positive heat flux into the control volume.

The thermal properties of biological materials change with temperature, and this is reflected in the measured values which appear in literature (Bowman et al 1975). In this work we assume that the thermal properties of the materials involved remain constant, which is true for small changes in temperature, like those appearing in the presented applications.

3.3. Some features of the computer program

The derivation of the discretization equations with the control-volume formulation and the consequent use of the weighted thermal conductivity in the calculations allow the handling of inhomogeneous bodies in a physically consistent manner. The program can take advantage of the non-uniform meshes used in the FDTD kernel and, thus, work with an acceptable discretization of arbitrarily shaped objects without waste in computer memory.

The most important feature of the program is that it can allow time scaling in the way suggested by Torres and Jecko (1997). The heating process in realistic situations lasts for seconds or minutes, whereas the electromagnetic steady state is reached within a time period smaller by many orders of magnitude in microwave frequencies. In order to simulate the much longer heating times, a way must be found to accelerate the heat diffusion process. It has been shown by Torres and Jecko (1997) that the multiplication of the dissipated power and the thermal conductivity of materials with a factor \( \alpha > 1 \) can make the time evolution of temperature faster by this factor, without altering its spatial distribution. Let us consider the case in which we would like to have an actual heating duration of \( t_{\text{heating}} \) to be simulated within the FDTD computation window of \( t_{\text{FDTD}} \). Then, we would choose a scaling factor of

\[
\alpha = \frac{t_{\text{heating}}}{t_{\text{FDTD}}}.
\]

The cost that one has to pay for the acceleration of the heat conduction process is the use of a smaller time step for the numerical simulation, i.e. the stable time step for heat conduction
becomes smaller by $\alpha$. Nevertheless, in most cases the stable time step of the electromagnetic simulation is smaller than the stable scaled time step of heat diffusion and can, therefore, guarantee numerical stability of the thermal simulation as well. The feature of time scaling allows for coupled electrothermal simulations with reduced computational times. Torres and
Figure 2. (a) Comparison of the analytical and numerical temperature distributions of a spherical heat source along its main diameter. (b) Distribution of the numerical error (relative to the analytical solution) on the plane that goes through the center of the spherical heat source.

Jecko (1997) have examined the influence of the scaling factor $\alpha$ on the results accuracy and have found that even for large values of the scaling factor ($\approx 10^{10}$) the relative error remained small ($<1 ^\circ C$). Although not used in the examples presented in this work, the capability to perform coupled electrothermal simulations reduces the simulation time by considering the temperature dependence of the complex Debye permittivity of the materials and employing the (FD)$^2$TD formulation (Torres and Jecko 1997).

The computer program works in the way shown in figure 1, performing both sequential and coupled electromagnetic and thermal simulations. The time points for the initiation and the duration of the simulations are determined by the user.

3.4. Validation of the heat conduction computer program

In order to validate the developed computer program, we have calculated a number of examples and compared the results with their analytical solutions. As expected, the error was higher for objects not conforming to the rectilinear mesh, i.e. spherical and oblique objects. Figure 2(a) shows the comparison between the analytical and the numerical solution of a spherical heat source. The sphere of 10 cm radius was discretized with a uniform grid of cubical cells of side 0.5 cm. The total computational domain consisted of $40 \times 40 \times 40$ cells. The material of the sphere simulated brain tissue with mass density $\rho = 1030 \text{ kg m}^{-3}$, thermal conductivity $k = 0.528 \text{ W m}^{-1}\text{C}^{-1}$ and specific heat $c = 3710 \text{ J kg}^{-1}\text{C}^{-1}$ (Bowman et al 1975). The initial temperature used was $37 ^\circ C$ whereas the ambient temperature was kept at $25 ^\circ C$ (Dirichlet boundary conditions). The heat generation rate inside the sphere was assumed uniform and equal to $5000 \text{ W m}^{-3}$, which corresponds to an SAR of about $4.85 \text{ mW g}^{-1}$. The time step used was $15.07 \text{ s}$ (the stable time step) for 200 steps, i.e. a total simulation time of approximately 50 min.

The error distribution on the plane that goes through the centre of the sphere is shown in figure 2(b). The maximum error of the calculated temperature distribution is $1.75\%$ and appears at the boundaries of the sphere, where the discretization errors are larger in this case.
4. Validation of SAR assessment with temperature probes

Thermal simulations can determine the error introduced by the mechanism of heat conduction when assessing experimentally the SAR with temperature probes and the equation

\[
\text{SAR} = \frac{c \Delta T}{\Delta t}
\]  

(2)

where \(c\) is the specific heat of the biomaterial and \(\Delta T\) is the temperature increase measured during time \(\Delta t\). The temperature rise can be computed with the developed software at any point inside the experimental volume, taking into account heat conduction for the time period \(\Delta t\). Thus, by employing equation (2) we can calculate the SAR that would be assessed by an ideal probe at that point. The term *ideal* is used here as opposed to the real temperature probes which not only might interact with the electromagnetic field but also measure the mean temperature of a discrete volume, introducing further errors in the experimental assessment of SAR. In this work we study only the effect of heat conduction on the SAR measurements.

4.1. Example I: Petri dish

The dosimetry of Petri dishes filled with different amounts of medium and exposed in a TEM cell in E-polarization (E-field parallel to the axis of the Petri dish) has been presented in Burkhardt et al (1996). The study, from which an analytical approximation was derived, was based on two numerical techniques, the finite integration technique (FIT) (Weiland 1990) and the generalized multipole technique (GMT) (Hafner and Bomholdt 1993). In the study an experimental validation was also performed with a temperature probe, but there was a significant deviation between the calculated and the measured results, especially in the centre of the dish. On the contrary, in an additional experimental validation performed by the same group and using a novel E-field probe designed for this kind of applications with a sensor size of only 1 mm (Poković et al 1997), the obtained data almost perfectly matched the numerical results. In the context of the present study, it was interesting to evaluate the error that is introduced by heat conduction when using a temperature probe to assess the SAR.

The SAR and temperature distributions were calculated with the new software. The dielectric properties of the materials were the same as in Burkhardt et al (1996) and the thermal properties used are given in table 1. The incident power density was assumed to be 240 mW cm\(^{-2}\) at 835 MHz. A non-uniform mesh was used in the calculations with the smallest voxel size being 0.6 \(\times\) 0.6 \(\times\) 0.4 mm\(^3\). The computational domain was truncated with second-order Mur’s absorbing boundary conditions. The stable time step for the FDTD was approximately 0.97 ps and the total simulation time was seven time periods of the signal, corresponding to a total of 8652 time steps.

The SAR distribution was calculated from the electric field along the main axis of the Petri dish from

\[
\text{SAR} = \frac{\sigma |E|_{\text{max}}^2}{2\rho}
\]  

(3)
Figure 3. Comparison between measured and calculated SAR distributions scaled for an exposure power density of 1 mW cm$^{-2}$. The distributions with heat conduction were calculated using equation (2) and the distributions from the FDTD method and the measurements were calculated using equation (3).

where $\sigma$ is the electric conductivity and $\rho$ the mass density of the suspension and $|E|_{\text{max}}$ is the norm of the maximum total steady-state electric field at a point. The numerical results for the SAR distribution are the same as in Burkhardt et al. (1996), because the FIT and the FDTD techniques lead essentially to the same numerical formulation.

The temperature was calculated with the new software, after electromagnetic steady state had been reached, for different exposure times. The longest measurement time which was simulated for this example was 20 s. Insulating boundary conditions were used for the heat conduction simulation. Using equation (2) the SAR distribution was then assessed. The results are summarized in figure 3.

It is interesting to note that even a measurement interval of 5 s causes errors of up to 95% for the prediction of the local SAR values at the centre of the dish with an ideal temperature probe. It can be observed that the SAR at the centre of the dish is overestimated and at the top and bottom of the dish it is underestimated. These results agree with the results of Pickard et al. (1999) who studied the exposure of a culture flask in a transverse electromagnetic mode exposure system. Although the numerically predicted SAR distribution is different in the two cases, they also found that in a 5 s measurement interval, an error of as much as 100% could be expected. In their work, Pickard et al. (1999) mention that they tested different values of the heat transfer coefficient at the upper surface of the medium but found no influence on the rate of temperature elevation at the cell layer during the first 30 s of heating. Therefore, the use of adiabatic boundaries in the case of the Petri dish is not expected to diminish the
validity of the presented results, since the medium height in the example presented here is about the double of the medium height used by Pickard et al (1999) in the culture flask. It can be deduced from the above that in the examined case the miniaturized E-field probe has the best overall performance compared to the temperature probe, whose accuracy is degraded to a large extent due to the effect of heat conduction. However, both methods, i.e. thermometric and electromagnetic, deviate from the FDTD calculations when measuring SAR at the ‘cell layer’ (z < 1 mm in figure 3). This fact calls for further research and development.

4.2. Example II: carousel exposure set-up

The carousel set-up and its dosimetry have been described in Burkhardt et al (1997). It is a partial body exposure set-up designed for testing in vivo the effects of wireless communications systems on the central nervous system (CNS). The detailed dosimetry in that study was carried out by the FIT tool MAFIA (CST GmbH, Büddingerstr. 2a, D-64289 Darmstadt, Germany) based on an MRI model of a rat. An exposure system which is based on the carousel and is called ‘the chamberette’ has recently appeared in the literature (Moros et al 1998). The experimental dosimetry of this latter exposure set-up has been presented in Moros et al (1999).

In the study by Burkhardt et al (1997) several parameters of the model have been varied, in order to assess the uncertainties of the calculations. The validation was conducted with thermal probes inserted in holders inside the trunk and the brain region of a rat cadaver. The system used was a Photronics 1450 HS Fiberoptical Multisensor System (Photronics, 142nd Avenue, 18 800 Woodinville, WA 98072, USA). In their discussion of uncertainties, the authors mentioned that due to poor sensitivity of this system (>10 mK s$^{-1}$) a measurement time of 60 s had to be chosen, even when the antenna input power was 82 W. They indicated that the effect of heat conduction might be significant at locations of high SAR gradients. It will be shown in the present study that such an effect can indeed become essential.
We have used a discretized MRI rat model (slightly different from the one used by Burkhardt et al. (1997)) placed inside the exposure set-up with its snout 3 cm away from the antenna. The discretization with a graded mesh resulted in about 1.1 million voxels. The smallest step in space was 0.5 mm. The physical properties of the rat model are shown in table 2. An initial temperature of 22°C and insulating boundary conditions have been assumed.

Figure 4 shows two calculated distributions of the SAR$^V$† along a line that goes through the brain of the rat. Both the SAR$^V$ distributions in figure 4 have been calculated with the developed computer program. One distribution results from the values of the electric field by using equation (3) and the other corresponds to the values which would be measured after 1 min with an ideal temperature probe. The latter distribution results from the calculated temperature elevation and equation (2). It can be seen that the SAR distribution which would be assessed with an ideal temperature probe deviates from the numerical one by as much as 170%.

The availability of both electromagnetic and thermal simulations makes it possible to determine the error introduced by the mechanism of heat conduction while assessing the SAR with temperature probes and the correction that is needed to overcome this error. A first approach to this was presented by Moros and Pickard (1999). In their work they calculated two ratios as a function of a dimensionless similarity variable, which depended on the measurement time, the thermal diffusivity of the medium and the half-width at half-maximum of the SAR distribution. The ratios were the SAR values predicted from the tangential and chord slopes

† SAR$^V$ denotes the volume related SAR, i.e. $\text{SAR}^V = \rho \times \text{SAR}$. 

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Figure 4. SAR$^V$ distribution along the line at $x = 70.25$ mm which goes through the rat’s brain. It can be seen how the SAR$^V$ distribution which is calculated numerically differs from the one that would be estimated using an ideal temperature probe and equation (2).
Electrothermal computations for NIR dosimetry

Figure 5. Variation of the correction factor along a line (see also figure 4) that goes through the rat’s brain for several measurement time intervals. (a) The specific heat of water is assumed for all tissues. (b) At each point the tissue specific heat is considered.

of the temporal temperature change against the true SAR. Here, we define a similar quantity, which we call the ‘correction factor’. It represents the quantity by which the SAR_{HC} value, assessed with the consideration of heat conduction, needs to be multiplied, in order to result in the value SAR_{FDTD}, calculated with a numerical electromagnetic tool:

$$\text{CF} = \frac{\text{SAR}_{FDTD}}{\text{SAR}_{HC}}.$$  \hfill (4)

The correction factor CF of equation (4) can be viewed as the reciprocal of $R_{chord}$ in Moros and Pickard (1999), when only two points in time are assumed and the SAR_{FDTD} is considered as the true SAR.

Apparentely, the correction factor CF as well depends on material thermal properties, measurement time and position, i.e. CF = f(D, x, y, z, t), where D is the medium thermal diffusivity. Figure 5 shows this variation for the points along a line that goes through the rat’s brain (see also figure 4). It can be seen (figure 5(a)) that when experimental SAR values are assessed with a constant specific heat, i.e. that of water, then there is likely to be an overestimation of the real SAR values (CF < 1). This result does not contradict the conclusion of Moros and Pickard (1999) that in a cylindrical geometry (corresponding to the rat’s head) there can never be an overestimation of SAR due to thermal diffusion. Here it is only pointed out that the knowledge of the right specific heat value is crucial for inferring SAR from temperature measurements. Besides, the conclusion of Moros and Pickard (1999) that SAR is more often thermometrically underestimated can be observed clearly in figure 5(b).

In this figure we take into account the specific heat of the measured tissue at each simulated measurement point. It can be seen that the three SAR peaks (one each in muscle, bone and brain) are underestimated with respect to the real values (CF > 1), since the process of heat conduction tends to equalize the temperature gradients.

The SAR peaks in the three tissue areas have a half-width at half-maximum ($\alpha$) which is \approx 6.6 mm in muscle, \approx 1.5 mm in bone and \approx 5 mm in brain (figure 4). The head of the rat as a whole can be considered to have cylindrical geometry. Following the approach of Moros and Pickard (1999), we find out that for this type of geometry and for the thermal properties of the respective tissues (table 2) the maximum measurement time to ensure an error of less than 10% due to heat diffusion is \approx 25 s for muscle and \approx 15 s for brain. These results agree
Figure 6. SAR$^V$ distribution in the middle sagittal plane of the rat model. (a) Calculated numerically. (b) Predicted with equation (2) for temperature measurements after 1 min of irradiation, with heat conduction taken into account.

quite well with the curves of figure 5(b), especially for the SAR maximum in muscle for which the CF reaches the value of 1.1 (10% error) at 24 s. The SAR maximum in bone falls in the category of sub-centimetre volumes ($\alpha < 3$ mm) for which Moros and Pickard (1999) claim that the measurement time has to be smaller than 12 s to achieve heat diffusion errors below 10%; this can be observed clearly in figure 5(b).

5. Determination of the highest thermal load

In the process of planning bioexperiments one difficult task is to decide on the exposure levels in terms of SAR. Especially for experiments which investigate the biological effects of low-level electromagnetic exposure it is important to be able to determine the exposure level above which thermal effects will definitely occur. This exposure level corresponds to the maximum expected temperature rise in the subjects, i.e. the highest delivered thermal load, for the calculation of which both electromagnetic and thermal simulations are necessary. When the electromagnetic problem is exactly defined, i.e. parameters like geometry, radiating structure, frequency, modulation scheme, power level, etc. do not vary, the maximum temperature rise can be calculated by excluding the heat exchange mechanisms (‘worst-case approach’) in the numerical model, i.e. by omitting convection and radiation.

An example is given here for the simulation of a 1 min irradiation of the MRI rat model in the carousel exposure set-up. The combined electromagnetic and thermal simulations gave the distributions shown in figure 6. The maximum temperature increase in the rat was 0.22°C. The spatial average values for the worst-case temperature elevation in the brain and the whole-body were 0.09°C and 0.01°C respectively.

Recently Moros et al (1999) have presented the dosimetry of an exposure set-up which is based on the carousel. According to their account a large rat of 520 g (which corresponds to our MRI rat model) would have an average SAR in brain tissue of 1.11 ± 0.18 (SD) W kg$^{-1}$ per W of input power when the head is at ‘middle’ position. In their work they have also assessed the thermal load on the animals during irradiation. They found that during a period of 20 min no appreciable temperature increase occurred for an irradiation at 1.5 W input power, i.e. about 1.65 W kg$^{-1}$ average SAR in brain tissue. In the example shown in figure 6(b) the irradiation was simulated for 1 min and for an average brain SAR of 5.2 W kg$^{-1}$, which can be achieved with a 240 mA antenna feedpoint current (rms value). If we scale the calculated whole-body average temperature rise for the time and the SAR of Moros et al (1999), the
expected temperature elevation for 20 min irradiation at 1.5 W of input power is 0.06 °C, which is indeed not appreciable. It should be noted that this calculation is very conservative due to the assumptions made (adiabatic boundaries, absence of blood perfusion, linear rise of temperature with time).

6. Conclusions

We have presented the integration of a heat conduction module into an FDTD electromagnetic solver to create a flexible computer program which can support (a) the study of complex biological bodies, and (b) longer heating durations with reduced computation times. The importance of such a tool in NIR dosimetry has been demonstrated by two applications: it has been shown that the assessment of the SAR distributions with less sensitive temperature probes suffers from significant errors due to heat diffusion. The results of this application agree well with the findings of previous studies (Moros and Pickard 1999, Pickard et al 1999). Moreover, it has been shown that the computer tool is capable of estimating the maximum thermal load conveyed to the subjects of experiments studying the biological effects of electromagnetic radiation. Work is currently in progress to integrate the full BHTE into the software for applications of RF risk analysis, although it can be used in this area also in its present form when only heat conduction is significant (see Taflove and Brodwin 1975, Bernardi et al 1998).

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