A Three-Dimensional Thermal and Electromagnetic Model of Whole Limb Heating with a MAPA

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Abstract—Previous studies by the authors have shown that if properly implemented, the Pennes assumptions can be applied to quantify bioheat transfer during extremity heating. Given its relative numerical simplicity and its ability to predict temperatures in thermoregulated tissue, the Pennes model of bioheat transfer was utilized in a three-dimensional thermal model of limb heating. While the arterial blood temperature was assumed to be radially uniform within a cross section of the limb, axial gradients in the arterial and venous blood temperatures were computed with this three-dimensional model. A realistically shaped, three-dimensional finite element model of a tumor-bearing human lower leg was constructed and was “attached” mathematically to the whole body thermal model of man described in previous studies by the authors. The central as well as local thermoregulatory feedback control mechanisms which determine blood perfusion to the various tissues and rate of evaporation by sweating were input into the limb model. In addition, the temperature of the arterial blood which feeds into the most proximal section of the lower leg was computed by the whole body thermal model. The variations in the shape of the tissues which comprise the limb were obtained from computerized tomography scans. Axial variations in the energy deposition patterns along the length of the limb exposed to a miniannular phased array (MAPA) applicator were also input into this model of limb heating. Results indicate that proper positioning of the limb relative to the MAPA is a significant factor in determining the effectiveness of the treatment. A patient-specific hyperthermia protocol can be designed using this coupled electromagnetic and thermal model.

INTRODUCTION

SEVERAL FACTORS must be considered when formulating a model of the thermal response of the human body to the heat sources imposed during a therapeutic hyperthermia treatment. Quantifying the effects of blood perfusion on tissue temperature is one of the most significant aspects of such a formulation. Thermoregulation of blood flow within both the skin and muscle tissues has been shown to be a primary method for shedding heat to the environment under conditions of heat stress [1]–[3] or retaining heat in the body during cold exposure [4]. As discussed in several papers by Charny and Levin [5]–[7],

there have been several approaches to modeling these bioheat and mass transfer phenomena.

Weinbaum *et al.* [8] have presented a model which relates the microcirculatory network geometry to an effective thermal conductivity. This effective thermal conductivity accounts for the enhancement of heat conduction in perfused tissue based upon the size and spacing of the branching countercurrent vessel pairs. This model assumes that the gradient of the mean tissue temperature in a region can be approximated by the gradient of a weighted mean of the temperatures of the blood in the arterioles and venules located in the tissue. As shown in Charny and Levin [7], this assumption is questionable under the conditions imposed by hyperthermia and even under some normothermic conditions.

Two recent theoretical studies by Charny *et al.* [6], [7] revealed that the traditional bioheat transfer model of Pennes [9] was able to predict tissue temperatures within a slab of heated tissue under hyperthermic conditions within 1/2 °C of those computed by a more complex model which accounted for the thermal effects of a branching countercurrent microcirculatory system. These results led to the conclusion that the thermal response of a more complex, three-dimensional model of a human extremity to hyperthermia could be effectively quantified with a properly implemented version of the Pennes model for bioheat transfer. More specifically, while we may assume that the arteriole blood temperature is uniform throughout a given cross section, we may not assume that the arterial blood temperature is uniform from cross section to cross section in the three-dimensional model.

Another important factor in the formulation of the limb model is the leg geometry. Human extremities have most often been modeled as a series of coaxial cylinders [10]–[15]. This approach is a reasonable first approximation. However, a series of computerized tomography (CT) scans reveal that the lower leg, while fairly cylindrical, consists of two eccentrically positioned bones, as well as at least one set of major arteries and veins embedded within the muscle tissue. By digitizing these CT scans, irregularly-shaped elements were built and combined to form the finite element model of the lower leg presented in this paper.

The radio frequency (RF) energy deposition patterns in a leg exposed to a MAPA are also important in determining tissue temperatures during hyperthermia. Due to the irregular shape of an extremity and the physical phenom-

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Fig. 1. Two computerized tomography (CT) sections of a human lower leg. These two sections are located 3 cm apart.

ena which induce current flow in a limb exposed to the MAPA, axial variations in the RF-heating pattern will depend on the leg geometry and its position relative to the MAPA applicator. As shown in Charny *et al.* [16], this effect may induce a major shift in the axial position of maximum energy deposition along the length of the leg. Thus the model presented in this paper includes this axial shift phenomenon. The empirical model presented in Charny *et al.* [16] was utilized to quantify this effect.

Finally, as shown in Charny and Levin [10], [11], the thermal responses of extremities to hyperthermia depend not only on local temperatures, but also on the whole body temperatures. Central as well as local thermoregulation affects the limb thermal response. In addition, the temperature of the arterial blood which feeds into the lower leg can be estimated with a whole body thermal model. It has been generally assumed that the temperature of the inlet arterial blood that feeds a heated region is constant at the normothermic body core temperature of 37°C throughout a hyperthermia treatment [17]–[21]. However, one important result of the simulations presented in Charny and Levin [10], [11] was that the temperature of the arterial blood that feeds the lower leg increases on the order of 2°C during a 1 h MAPA treatment. This effect is due to the systemic heating that occurs during regional hyperthermia treatments. Thus, it is necessary to attach a whole body thermal model to the detailed limb model so that this systemic effect can be included in the simulations of MAPA heating.

The three-dimensional model that is presented in this paper therefore represents a combination of the important physical and physiological phenomena that govern heat flow in a tumor-bearing limb during hyperthermia. These effects, which were investigated individually in previous studies can now be combined in order to confidently model the treatment of tumors within human limbs during hyperthermia. This model can be used to evaluate the importance of the various physical and physiological effects discussed above on normal and tumor tissue temperatures,

thereby providing a valuable tool for hyperthermia treatment planning.

DESCRIPTION OF THE THREE-DIMENSIONAL FINITE ELEMENT MODEL

A. Limb Geometry

A series of CT scans of a nontumor-bearing lower leg was obtained from the Radiation Oncology Branch of the National Cancer Institute. The lower leg, which was 30 cm long from the base of the knee cap ($z = 0$ cm) to the proximal edge of the ankle ($z = 30$ cm), was viewed as a series of thirty 1.0 cm thick cross sections. The 30 images were transferred to paper (hardcopy) sheets. Eleven of these images were then digitized using a Tektronix® 4957 digitizing tablet. The eleven images were spaced 3 cm apart, i.e., at $z = 0, 3, 6, \dots, 30$ cm. For each of these eleven cross sections, outlines were made of the various tissue boundaries which define the leg: the tibia, the marrow located within the compact bone of the tibia, the fibula, the muscle, the fat, and the skin. The skin was subdivided into two sections, an inner and an outer layer. CT scans of two of the leg's cross sections located near the central plane of the lower leg, at $z = 18$ and $z = 21$ cm, are shown in Fig. 1.

The use of CT scans in modeling the limb geometry allows for an accurate, patient-specific treatment simulation. A simpler approach would be to obtain cross-sectional information from an anatomy atlas, along with measurements of the patient's bone length and the relative position of each cross section. In some cases, however, such as when the limb sarcoma is large, the presence of a tumor in the limb can significantly alter the geometry of a portion of the limb relative to the normal shape. In these cases, some error would be introduced into the heat transfer model via this approximation of the leg geometry. This error would be especially significant in terms of computing the electromagnetic energy deposition pattern in the limb. For these reasons, and because CT scans are readily

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Fig. 2. A three-dimensional finite element "axial section" of the lower leg. These finite elements were constructed by digitizing the tissue interfaces in Fig. 1 and subsequently connecting the grids to form three-dimensional elements.

Fig. 3. A view of the 1038 three-dimensional finite elements that comprise the lower leg. Axial sections #1 and #10 are located at the proximal and distal ends of the tibia, respectively. A 100 mL soft tissue sarcoma is located in axial sections #4 and #5.

available for patients receiving these hyperthermia treatments, the patient-specific geometry is utilized rather than a more general shape.

The digitized grid points were input into the PATRAN II® finite element program. Two-dimensional elements were formed from the input grid points at each of the eleven digitized cross sections. Three-dimensional hexagonal and tetrahedral elements were built by connecting the two-dimensional elements at adjacent sections. Fig. 2 illustrates the "axial section" of tissue that results from this construction procedure for the two planar images shown in Fig. 1. A 100 mL soft tissue sarcoma was modeled by replacing several adjacent muscle elements between $z = 9$ cm and $z = 12$ cm in the lower leg with tumor elements.

A total of 1038 tissue elements, consisting of 1255 nodes, were used to model the lower leg. These elements were subdivided into the ten axial sections shown in Fig.

3. The most proximal and distal axial sections are referred to as sections 1 and 10, respectively.

One major artery and one major vein were assumed to supply and drain blood to the lower leg. These vessels were positioned according to anatomical references [22]. The blood elements were constructed in the finite element model with rectangular channels having a hydraulic diameter equivalent to the diameter of the vessels *in situ*. Based upon data from the literature, the diameters of the artery and vein at the proximal end of the lower leg were 0.35 and 0.70 cm, respectively, while their center-to-center spacing at this axial location was 0.525 cm [23]. These vessels were assumed to taper along the length of the lower leg in a linear manner such that their diameters and spacing at the distal end of the limb were one-half those at the proximal end [24]. Ten arterial and ten venous blood elements were constructed in this limb model.

The foot attached to the heated lower leg was modeled

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by extending the elements in the lower leg axial section number 10 in the distal direction such that the total foot volume was equivalent to that of the foot in the whole body model of Charny and Levin [10], [11]. This foot segment was subdivided axially into two segments, each 11 cm in length.

B. Tissue Properties

The elements which comprised the lower leg were characterized according to their tissue type. Only three of the tissues were considered to be perfused by blood: the muscle, the inner skin, and the marrow. The others were assumed to have no significant blood perfusion. Similarly, the basal metabolic rates in these unperfused tissues were neglected. Metabolic heating rates and basal perfusion rates in the muscle, skin, and marrow were obtained from the whole body model input data listed in Table I of Charny and Levin [10]. Each tissue was also characterized by a thermal conductivity, specific heat, and density, also listed in Table I of [10]. The data in this table, which were obtained from various sources in the literature, describe the physical properties of human bone, muscle, fat, skin, and blood, as well as brain, thoracic, and abdomen core tissues.

The tumor elements were assumed to have the same physical properties as the normal muscle. However, their basal blood flow rate was elevated at a level ten times that of normal muscle, and this tissue was not subject to the thermoregulatory mechanism observed in normal muscle [27]. The degree to which tumor blood flow is elevated relative to the normal muscle tissue varies significantly, depending on the tumor type. In the study by Waterman *et al.* [27], a factor of ten was typical of some tumors. In the present study, this value was used to simulate blood flow in the tumor periphery, while the tumor core was assumed to be necrotic and thus had zero blood flow. Since the tumor periphery represents the most viable portion of a sarcoma, it is important to heat this region to the therapeutic temperature level. A lower value for blood flow in the tumor periphery would most likely result in higher temperatures during a hyperthermia simulation in this region due to the cooling effect of perfusion. Thus, modeling the tumor with a relatively high blood flow rate will tend to underestimate the degree of heating in this critical region. This underestimation provides a reasonable guideline for evaluating the potential success of the simulated hyperthermia treatment, which depends upon heating a tumor to a minimum therapeutic temperature near 43°C.

Normal muscle elements were subject to the threshold thermoregulation mechanism first observed by Sekins *et al.* [2] and later modeled by Charny *et al.* [6], [7]. This mechanism keeps the local rate of capillary blood flow in the muscle at a basal level as long as the local temperature is below a threshold value between 42–43°C. Once this threshold is exceeded, the thermoregulatory mechanism is turned on and the local muscle capillary perfusion rate increases according to a simple linear dependence with

the local temperature. This increase is not instantaneous, but subject to a first order time constant delay. In addition, the local muscle perfusion rate is limited to a maximum value, fixed at 25 times the basal rate in the present study.

Similarly, blood perfusion in the inner skin elements was computed according to both local and whole body temperatures. Based upon the work of Stolwijk and Hardy [3], the local skin perfusion rate depends upon the local skin temperature, as well as a whole body weighted mean skin temperature. A set of empirical relationships between these two skin temperatures and the local skin perfusion rate is presented in [3] and was also used by Charny and Levin [10], [11] in their whole body thermal model of man during hyperthermia.

In the present study, whole body temperatures were determined by mathematically linking the lower leg finite element model with the whole body thermal model of Charny and Levin [10], [11]. This whole body model subdivides the remainder of human body into 17 multilayered tissue segments that are connected by arterial and venous blood segments. Mass and energy balance equations are used to simulate time and spatial variations in tissue and blood temperatures throughout the body during a hyperthermia treatment. A mathematical description of the human thermoregulatory system is included in this model to account for the effects of sweating and vasodilation on blood and tissue temperatures during this procedure.

The mass flow rates of the arterial and venous blood at the various axial locations were computed via a mass balance procedure. The mass flow rate at a given location was determined by subtracting the blood perfusion to the tissues at that location from the mass flow rate in the vessels at the adjacent proximal position.

C. Energy Balance Equations

The energy balance equations used to compute temperatures at each tissue node i ($i = 1, 1255$) in the finite element model was

tissue

$$\rho_i c_i \frac{\partial T_i}{\partial t} = k_i \nabla^2 T_i + Q_{\text{met},i} + Q_{\text{rf},i} - \omega_i (\rho c)_{bl} (T_i - T_{a,i}(z)) \quad (1)$$

where bl represents the properties of blood, and $T_{a,i}$ is the temperature of arterial blood that perfuses the tissue represented by node i .

The single term on the left-hand side of (1) represents the thermal capacitance of the tissue, while the right-hand side contains the heat conduction and three heat source terms. The first two heat source terms represent the heating effect, in W/m^3 tissue, of the metabolism (met) and hyperthermia (RF, for radio frequency) sources, respectively. The third heat source term in (1) represents the heating effect of capillary perfusion, which was originally described in the Pennes bioheat equation. Note that the

perfusion heat source is proportional to the rate of blood perfusion in the tissue ω and the local temperature difference between the tissue and arterial blood ($T_i - T_{a,i}$).

It is significant that $T_{a,i}$ is a function of axial position along the length of the lower leg but independent of the radial position at a given axial location. This latter assumption, based upon the Pennes model, was investigated in Charny *et al.* [6],[7]. These two studies concluded that heat transfer in muscle tissue perfused by a branching, countercurrent arteriovenous network could be modeled by the Pennes bioheat equation under conditions of basal and moderately elevated blood flow rates. This conclusion was reached by comparing predictions of the Pennes equation with those of a more rigorous three-equation model from which the Pennes equation can be derived. Calculations of tissue temperature profiles under both steady-state normothermic conditions and transient hyperthermic conditions were utilized in these bioheat transfer analyses.

The energy balance equation for the arterial blood element j ($j = 1, 10$) was

artery

$$V_{A,j}(\rho c)_{bl} \frac{\partial T_{Aj}}{\partial t} + m_{A,j} c_{bl} \frac{\partial T_{Aj}}{\partial z} = V_{A,j}(Q_{\text{met},A,j} + Q_{\text{RF},A,j}) \quad (2)$$

and similarly for the venous element k ($k = 1, 10$)

vein

$$\begin{aligned} V_{v,k}(\rho c)_{bl} \frac{\partial T_{vk}}{\partial t} - m_{k} c_{bl} \frac{\partial T_{vk}}{\partial z} \\ = V_{v,k}(Q_{\text{met},v,k} + Q_{\text{RF},v,k}) \\ + \sum V_i(\rho c)_{bl} \omega_i (T_i - T_{v,k}(z)). \end{aligned} \quad (3)$$

The artery and vein heat balance equations are similar as they both have a thermal capacitance and axial bulk flow term on their left-hand sides. Note that the flow rate of blood in the vein is in the negative z -direction. Equations (2) and (3) also have volumetric metabolic and hyperthermic heat source terms on their right-hand sides similar to those in the tissue heat balance equation.

The last term on the right-hand side of (3) accounts for the heat flow into the venous blood due to the complete thermal equilibration that takes place in the microcirculation between the blood in the venules and the tissue elements which are perfused by these microvessels. The summation in (3) is formed within a given axial section of the leg. Thus, all of the tissue elements i positioned within a particular axial section are assumed to exchange heat with the venous element that flows within this "disk" via this equilibration mechanism. The terms contained in (1)–(3) are analogous to those which constituted the tissue and blood energy balances for the whole body thermal model in Charny and Levin [11].

Equation (1) was subject to a no-flux boundary condition in the axial direction at position $z = 0$ cm. The lower

leg boundary at the outer skin layer was subject to a convective boundary condition with either ambient air or some other medium with which the skin was in contact. Similarly, the boundary at the most distal edge of the foot was governed by a convective boundary condition with the ambient air. This boundary condition may be expressed as

$$-k_i \frac{\partial T_i}{\partial n} = h_s(T_i - T_{\text{amb}}). \quad (4)$$

The inlet arterial blood temperature, i.e., T_A at $z = 0$, was determined by mathematically linking the lower leg finite element model to the whole body thermal model of Charny and Levin [11]. This procedure is described below.

While the heated lower leg and foot were modeled with the finite element model described above, the remainder of the whole body was represented by seventeen lumped body segments, each consisting of four tissue and two blood compartments. These whole body tissue and blood compartments were transformed into a series of nodes that were characterized by the same volumes, physical properties, basal metabolic and perfusion rates, and physiological characteristics as their corresponding elements in the original whole body model described in Charny and Levin [11]. Blood perfusion rates to the skin and muscle nodes were subject to the thermoregulatory control mechanisms described in Charny and Levin [11]. Similarly, the control of heat loss from the skin nodes by evaporation was subject to the sweating model described in Charny and Levin [11].

The whole body node that represented the arterial blood compartment in the upper leg of the heated limb was connected to the inlet arterial blood element of the finite element lower leg model with an "advective" flow element. In the same manner, the outlet venous blood element from the finite element lower leg model was advectively attached to the whole body node that represented the venous blood compartment in the upper leg. The energy balance equations utilized to compute the temperatures of the whole body tissue and blood nodes were those listed in (3)–(9) of Charny and Levin [11].

Coupling these two models provided a reasonable method by which to compute the inlet arterial temperature into the lower leg during a hyperthermia treatment. As shown in the simulations presented in Charny and Levin [10], [11], the value of the arterial feed blood can increase by 1–2°C over the course of a 1 h hyperthermia treatment. The degree of systemic heating which occurred in the whole body under these conditions of heat stress was thus used to determine the magnitude and time course of any change in the lower leg inlet arterial temperature.

SIMULATION RESULTS

A. Steady-State (Normothermic) Simulations

Steady-state simulations were performed under basal, or resting conditions, with the ambient air temperature set

at 25°C and air velocity equal to 0.1 m/s. The steady-state temperatures at all 1255 nodes in the lower leg were reduced to a series of mean tissue temperatures at each axial section via a volume-weighting method. For a given tissue n at a particular z -location, the mean tissue temperature was computed by multiplying the temperature of each node i by the volume associated with that particular tissue node:

$$\langle T \rangle_n(z) = \frac{\sum_{i=1}^N V_{n,i}(z) T_{n,i}(z)}{\sum_{i=1}^N V_{n,i}(z)} \quad (5)$$

where N is the total number of nodes of tissue n in the axial section being examined. These mean steady-state temperatures in each of the seven tissues at each axial section of the lower leg are shown in Fig. 4. The axial variation in the temperatures of the blood in the major artery and vein are also shown in Fig. 4.

The normothermic, steady-state temperature distribution was used as the initial condition for the simulations of hyperthermia in the lower leg which are presented in the next section. Similarly, the steady-state whole body tissue and blood temperatures computed in this simulation were assumed to be the initial temperatures for these nodes prior to the imposition of hyperthermic conditions. This simulation was solved using approximately 12 CPU hours on a VAX-11/750 computer.

B. Transient (Hyperthermic) Simulations

The RF-energy deposition pattern into the lower leg was computed using the "axial shift" relationship presented in Charny *et al.* [16]. The axial variations in RF-energy deposition within the lower leg positioned so that the tumor was located within the central plane of the MAPA (SAR distribution #1) are shown in Fig. 5. This is the position most commonly used by clinicians who treat limb sarcomas with the MAPA. Within a given axial section, the deposited RF-energy was subdivided into the various tissue layers in that section based upon the electrical conductivity of the tissue. This method was also used in the whole body model simulations of MAPA heating [10], [11]. A total of 150 W was deposited into the lower leg. As in the simulations presented in Charny and Levin [10], [11], "aberrant" or nonlocal RF-energy deposition outside of the lower leg was neglected.

The lower leg was assumed to be surrounded by a water-filled bolus, thus preventing any heat loss by evaporation from the lower leg. The water bolus was considered to be at room temperature (25°C). The ambient air conditions were constant at 25°C and 0.1 m/s. Both the whole body skin and muscle nodes and the lower leg skin and tissue elements were subject to the thermoregulatory control mechanisms described previously. The blood flow in the tumor elements was fixed at a rate equal to ten times that of normal muscle. Approximately 20 CPU min were required to solve this numerical model for each minute of simulated time on a VAX-11/750 computer.

Steady-State (Normothermic) Results

Fig. 4. Mean tissue temperatures in the lower leg in each of the ten axial sections under normothermic conditions. Axial variations in the arterial and venous blood temperatures are also shown.

Fig. 5. Axial variations in specific absorption rate (SAR) in the lower leg exposed to a miniannular phased array (MAPA) hyperthermia applicator as predicted by the electromagnetic model of Charny *et al.* [16]. The leg is assumed to be positioned so that the tumor is located at the central plane of the MAPA.

Lower leg muscle and tumor temperatures at various times during the hyperthermia simulation were reduced to mean temperatures according to the volume-weighting method described by (5) above. These results are shown in Fig. 6. The volume-weighted mean bone temperatures in the heated lower leg are presented in Fig. 7. Figs. 8 and 9 illustrate the thermoregulatory blood flow responses of a single muscle and skin node in the central region of the lower leg, respectively. These responses were typical of those for each tissue during the hyperthermia simulation.

DISCUSSION

The normothermic temperature profiles displayed in Fig. 4 reveal the effects of several important physiological phenomena. First, the tumor is at a higher temperature

Fig. 6. Transient mean muscle and tumor temperatures in the lower leg subject to the SAR pattern shown in Fig. 5. Temperatures are shown at times 0, 3, 10, and 60 min.

Fig. 7. Transient mean bone temperatures in the lower leg subject to the SAR pattern of Fig. 5. Temperatures are plotted for the time 0, 3, 10, and 60 min.

Fig. 8. The blood flow responses of a muscle and tumor element in the central portion of the lower leg during the simulation of hyperthermia. Note that the tumor blood flow is independent of temperature.

Fig. 9. The blood flow response of a skin element in the central portion of the lower leg during the hyperthermia simulation.

under these resting conditions than the surrounding normal tissue. This result, also seen in the whole body simulations [11], is due to the elevated blood perfusion rate assumed for the tumor periphery. The warm arterial blood tends to heat the tissue which it perfuses. Under these conditions, the surrounding normal muscle tissue is heated to a lesser degree than the tumor due to its lower rate of blood perfusion.

Another effect of this relatively high basal tumor perfusion rate can be seen in the axial temperature profile of the venous blood. The blood flowing in the major vein is significantly heated as it passes through the axial sections containing the tumor (numbers 4 and 5), as seen in Fig. 4. Blood in the microcirculation is assumed to be thermally equilibrated with the tissue it perfuses. Therefore, the microcirculatory blood exiting the tumor in the venules will be significantly heated relative to that exiting the normal muscle tissue and skin. This warm blood adds a significant amount of heat to the venous pool as it drains into the major vein (see Fig. 4).

It is apparent from these studies that by attaching a whole body model of man to a realistically-shaped model of a lower leg, the thermal response of a limb to hyperthermia can be realistically evaluated. Based upon the results presented in Fig. 6, a soft tissue sarcoma in the lower leg can be heated to the temperature levels considered therapeutically effective. As seen in the whole body MAPA heating simulations [10], [11], the normal muscle tissue tends to autoregulate to a level below that of the tumor. The tumor, however, is not subject to this thermoregulatory mechanism and will be heated to therapeutic levels between 43–45°C providing that RF-energy can be focused in the tumor.

Fig. 5 illustrates the significance of leg position on axial SAR variations. In this case, the leg is assumed to be positioned in the usual clinical manner so that the tumor is located within the central plane of the MAPA. However, the effect of leg shape on the axial SAR variations, as detailed in Charny *et al.* [16], is apparent. The tumor,

located in axial sections 4 and 5, receives less RF-energy than the normal tissue located in axial sections 6 through 8.

Fig. 6 reveals that even under these less than ideal conditions the tumor is still preferentially heated relative to the normal surrounding tissue. Ideally, the tumor should be positioned within the MAPA applicator so that this region receives the maximum RF-energy in the entire limb. In axial sections 4 and 5, however, the tumor is still heated to therapeutic temperature levels while sparing the surrounding normal tissue even though the limb is not optimally positioned. A simulation in which the leg is positioned differently is described below.

It is important to note that the muscle tissue located 3–12 cm away from the tumor, in axial sections 6 through 8, is significantly heated. This normal tissue is not spared from receiving a large thermal dose due to the large amount of RF-energy deposited in this region. The normal muscle thermoregulatory mechanism can act to remove a significant amount of this large heat load, but only to a certain extent.

Similarly, the bone temperatures in axial positions 4 through 8 during this simulation are quite high, as seen in Fig. 7. These high bone temperatures are caused mainly by the flow of heat by conduction into this tissue from the surrounding heated muscle. This result is important as bone temperatures above 42°C may be intolerable for patients undergoing this treatment [25].

The muscle and skin blood flow responses illustrated in Figs. 8 and 9, respectively, are similar to those observed in the lower leg heating simulations presented in Charny and Levin [11]. Muscle blood perfusion increases by an order of magnitude during this simulated treatment. The muscle temperature undergoes a single oscillation, in the same manner that has been observed in dog limbs during constant power hyperthermia [26]. The lower leg skin blood flow also increases by an order of magnitude on the same time scale as that predicted by the simpler model of lower leg heating in Charny and Levin [11]. As reported by Waterman *et al.* [27], the tumor blood flow rate is fixed at a constant level equal to ten times the blood perfusion rate of the surrounding normal muscle tissue.

A second simulation was performed in which the leg was positioned so that the region of the leg that contained the tumor received the greatest amount of RF-energy deposition. The axial shift relationship outlined in Charny *et al.* [16] was implemented so that the region of maximum energy deposition coincided with the location of the tumor. This SAR distribution (#2) is shown in Fig. 10. In this case, the leg was positioned with the MAPA central plane located 6 cm proximal to the central plane of the tumor.

As evident by the temperature profiles displayed in Fig. 11, this alternate leg position resulted in a more localized heat treatment to the tumor. The tumor temperature was greater than that of the normal muscle tissue throughout the leg. Thus, by varying the position of the leg relative to this type of applicator, the RF-energy can be focused

Fig. 10. The SAR pattern that results by positioning the tumor 6 cm distal from the MAPA central plane, as predicted by the same model utilized in Fig. 5. Note that the region of maximum SAR is located in axial sections #4 and #5, where the tumor is situated.

Fig. 11. The mean muscle and tumor temperatures during a simulation of hyperthermia that implemented the SAR distribution shown in Fig. 10. Temperatures are shown at 0, 3, 10, and 60 min of simulated time.

into the tumor and normal muscle tissue can be spared from the deleterious effects of these large heat sources. It must be stressed that nonoptimal positioning of the limb within the applicator can result in potentially damaging thermal effects in the normal tissue. However, due to its low blood perfusion rate, the bone within the cross-section receiving the maximum amount of energy deposition will still be significantly heated in either leg/MAPA positioning scheme. Based on this study, patient tolerance to bone pain might be a limiting factor in maximizing the effectiveness of a MAPA treatment.

While detailed experimental data are not available for comparison with the predictions of this model, the simulated results are in good qualitative agreement with clinical observations. Due to patient tolerance limitations, very few temperatures can be invasively monitored in the clinical setting. However, based upon clinical data com-

piled at the Duke University Medical Center [28], 1 h treatments with the MAPA generally yield tumor temperatures between 41–45°C, depending on the specific patient and power level used during the treatment. While the success of these treatments in the treatment of cancer is still being evaluated, it is clear that the MAPA hyperthermia applicator has been successful in heating limb tumors to what are considered therapeutic temperature levels.

CONCLUSIONS

The simulations presented in this paper revealed that a realistically shaped three-dimensional model can be implemented to predict the thermal response of a tumor-bearing lower limb during hyperthermia. The three-dimensional model, which incorporated the whole body model of thermoregulation described in Charny and Levin [11], as well as the axial shift theory presented in Charny *et al.* [16], predicted that a tumor can be preferentially heated relative to the surrounding normal tissue. However, the position of the limb within the MAPA applicator is important.

The first study presented in this paper assumed that the lower leg was positioned such that the tumor was located at the central plane of the MAPA. This type of positioning is typical of that utilized in the clinical setting. Under these conditions, the axial shift theory predicts that more RF-energy will be deposited distal to the tumor than in the limb region containing the tumor itself. Given this electromagnetic heat source distribution, the three-dimensional model predicted that there would be significant heating of normal muscle and bone tissue in these distal regions. However, when the leg position was shifted axially by 6 cm, the electromagnetic axial shift theory from Charny *et al.* [16] predicted that the RF-energy deposition was at a maximum in the region containing the tumor. In this case, the tumor received the largest thermal dose of any tissue region in the lower leg, while the normal muscle temperature reached a steady-state level no greater than 42°C throughout the limb. A patient-specific hyperthermia treatment can therefore be planned and optimized to some extent based upon the geometry of the heated limb with the aid of this axial shift relationship.

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