Radio-Frequency Ablation in a Realistic Reconstructed Hepatic Tissue

This study uses a reconstructed vascular geometry to evaluate the thermal response of tissue during a three-dimensional radiofrequency (rf) tumor ablation. MRI images of a sectioned liver tissue containing arterial vessels are processed and converted into a finite-element mesh. A rf heat source in the form of a spherically symmetric Gaussian distribution, fit from a previously computed profile, is employed. Convective cooling within large blood vessels is treated using direct physical modeling of the heat and momentum transfer within the vessel. Calculations of temperature rise and thermal dose are performed for transient rf procedures in cases where the tumor is located at three different locations near the bifurcation point of a reconstructed artery. Results demonstrate a significant dependence of tissue temperature profile on the reconstructed vasculature and the tumor location. Heat convection through the arteries reduced the steady-state temperature rise, relative to the no-flow case, by up to 70% in the targeted volume. Blood flow also reduced the thermal dose value, which quantifies the extent of cell damage, from ~3600 min, for the no-flow condition, to 10 min for basal flow (13.8 cm/s). Reduction of thermal dose below the threshold value of 240 min indicates ablation procedures that may inadequately elevate the temperature in some regions, thereby permitting possible tumor recurrence. These variations are caused by vasculature tortuosity that are patient specific and can be captured only by the reconstruction of the realistic geometry. [DOI: 10.1115/1.2720912]

Introduction

Radio-frequency (rf) ablation is a minimally invasive procedure that has the potential for widespread use in hepatic cancer therapy. In the procedure, rf current is applied to the tissue, resulting in the conversion of electrical energy into heat with the intent of tumor necrosis [1–3]. Temperatures in the range of 50–60°C can start the process of denaturation in minutes [4]. Treatment sessions are generally 10–30 min in duration and produce spherical necrotic regions that are 3–5.5 cm in diameter [2].

Mathematical models are valuable for predicting the temperature rise within the organ during rf ablation, thereby enhancing the likelihood of developing an improved protocol for tumor ablation with minimal damage to surrounding tissue. Models can also be used to predict the optimal setting of operational parameters, such as probe geometry, placement, and power level, for a given tumor location and pathophysiology.

Previous models [5,6] for studying hepatic tumor ablation have employed different forms of the Pennes [7] bio-heat equation. In the Pennes approach, heat transfer due to blood flow is modeled using a nondirectional lumped heat sink proportional to the difference in temperature between the local tissue and the blood temperature. The proportionality constant, similar to heat-transfer coefficients, is typically assumed to be constant throughout the organ. While such an approach has proven useful in a wide variety of applications, it does not accurately model the convective effect of blood flow near a large vessel [6,8]. A number of theories other than the Pennes bio-heat equation have been put forward to assess the heat energy convected through arteries [9–13].

One possible alternative to the Pennes bio-heat equation is direct physical modeling of the heat and momentum transfer within the organ, including convection through large vessels [14]. In our approach, perfusion in the tissue outside of the large vessels is treated with the lumped-parameter approach, while the vector equations describing momentum and heat transfer are solved within the vessel domain. Direct physical modeling requires knowledge of the relevant vasculature, which may be acquired from medical image data. Acquisition of the vasculature geometry, and use of vectorized form of transport equations within the acquired geometry, significantly increases the complexity and computational time of the rf calculation. The benefit of the increased complexity is the capability of incorporating patient-specific information into the calculation. The technique therefore represents a first step toward optimizing an ablation procedure for a specific patient.

The present study analyzes rf heating in the case of a tumor located near the bifurcation point of a hepatic artery. The arterial geometry is reconstructed from MRI images of a porcine liver. To the authors’ knowledge, this is the first study to analyze rf ablation using direct physical modeling of the fluid flow and heat transfer in a reconstructed, three-dimensional hepatic geometry.

This study focuses on the single vessel, to quantify the effect of arterial blood flow on a nearby ablated region. In principle, other large vessels resolvable (e.g., a portal vein) in the medical images may be incorporated, at the expense of increased computation time. A rf heat source having a Gaussian energy distribution, ob-
Fig. 1 MRI images of the excised porcine liver cross section shown in (A) and (B), provided by Jan Johannessen (Food and Drug Administration, Rockville, MD)

tained from a curve fit to a previously analyzed profile, is employed [15]. To study the range of influence of the blood flow through the bifurcated artery on tissue heating, different tumor locations are considered. The governing equations are solved in the vessel and tissue domains using the finite-element (FE) method.

Temperature rise and thermal dosage, calculated from our FE model, are analyzed to determine the extent to which the efficacy of the ablation procedure is decreased due to cooling by large vessels. Asymmetries of the temperature field, caused by the vector nature of blood flow, are also evaluated.

Methodology

Sections discussed below provide the detail methodology of: (1) image-reconstruction protocol; (2) relevant equations of transport modeling in large vessels and perfused tissue; and (3) heat-source modeling.

Image Reconstruction. A total of 128 images spaced 625 μm apart are obtained from an excised porcine liver segment in a small-bore (45 cm) MRI system, controlled by a Tecmag Apollo single channel console running NT-NMR (Windows NT- Nuclear Magnetic Resonance) software. Three-dimensional reconstruction of the geometry is achieved by a region growing technique from the segmented MR scans [15]. In order to resolve the geometry of the arteries and surrounding tissue, images are segmented by automatic upper and lower threshold values, which are based on image contrast. All gray colored pixels between the threshold values are treated as hepatic tissue; thus, identifying the vasculature in the liver. Figure 1 shows the MR scanned image sections near the inlet (Fig. 1(A)) and after the arterial bifurcation (Fig. 1(B)).

An image reconstruction algorithm [16] is used to define the regions to be processed and to display the segmentation result. The number of triangular elements on the reconstructed surface determines the quality of the reconstruction; more elements indicate better resolution. The grey value interpolation method is used to interpolate the images and generate the three-dimensional triangular mesh. After checking the smoothness of the three-dimensional geometry, polylines with different colors are created to distinguish the segmented arteries and tissue surfaces. Subsequently, surface and curve fit to the polylines are incorporated and an IGES format of the geometry is imported for automatic, quadratic, and tetrahedral mesh generation [17]. Tetrahedral mesh is selected instead of hexahedral mesh since: (1) geometric decomposition, a time consuming preprocessing process, can be minimized; and (2) it produced a better quality mesh, typically suited for reconstructed geometry, with skewness of less than 0.82 and an aspect ratio less than 10.

Accuracy of the temperature rise for the tetrahedral element was checked for no flow condition. Temperature rise, obtained from the realistic geometry having a tetrahedral element, was compared with a simplified geometry, without any vessel (thus, avoiding volume decomposition) and no flow, with hexahedral elements. Temperature rise obtained from both the models was within a few percent, thereby justifying the use of the tetrahedral elements.

The excised liver tissue surrounding the reconstructed arterial wall is considered to be of cylindrical shape (Fig. 2) with a height of 80 mm and diameter of 170 mm, containing about 500,000 elements. The typical average diameter of the parent artery is 1.5 cm. The diameter ratio of the outer cylinder and the artery is above 10. The dimension for the outer cylinder was chosen to ensure that the boundary conditions have no effect on the temperature rise inside the tissue volume.

Figure 2(E) shows hypothetical tumor locations to be ablated using the rf energy. Three different tumor locations are chosen to study the impact of realistic vasculature on the tumor temperature distribution. In locations 1 and 2, the tumor center is positioned 8.5 mm and 12 mm from the artery bifurcation region, respectively. In location 3, the tumor is near the main branch of the artery with its center ~8 mm from the main artery wall. The tumor is spherical and 1.4 cm in diameter. The rf heat source, described below, is centered at the middle of the tumor.

Governing Equations. In principle the equations governing transport of momentum, heat, and electromagnetic energy are coupled, due to the temperature dependence of material properties such as electrical conductivity [18], viscosity, and thermal diffusivity. In this initial study, however, all material properties are assumed to be constant. The momentum and electric fields may then be determined independently of the temperature, and then incorporated into the energy equation to determine the temperature field.
Tissue Model. For this study, the hepatic tissue is modeled as a
solid region where the perfusion is treated with a lumped param-
eter approach [7]. The following energy equation for tissue media
is used to determine the temperature distribution
\[
\rho c_p \frac{\partial T}{\partial t} = \frac{\partial}{\partial x_i} \left( k \frac{\partial T}{\partial x_i} \right) + \omega \rho_b c_{bl}(T_{bl} - T) + q_s
\] (1)

Here \( T \) is the tissue temperature; \( c_p \) is the tissue heat capacity; \( \rho \) is
the tissue density; \( k \) is the thermal conductivity; \( q_s \) is the spatial
dependent Gaussian heat source described in the later section; \( T_{bl} \)
is the arterial blood temperature; \( \rho_b \) is blood density; \( \omega \) is blood
perfusion; and \( c_{bl} \) is the blood heat capacity.

Arterial (Continuum Blood) Model. The conservation of
mass within the vessel is expressed by the continuity relation
\[
\rho \frac{\partial u_i}{\partial x_i} = 0
\] (2)
where \( u \) is the fluid velocity.

The momentum equation is expressed by the Navier–Stokes
equation
\[
\rho u_i \frac{\partial u_j}{\partial x_j} = \frac{\partial}{\partial x_j} \left( \mu \frac{\partial u_i}{\partial x_j} \right) + \rho g_i + f_i
\] (3)
Here \( i, j = 1, 2, 3 \) for three dimensional flows; \( \partial u_i/\partial x_j \) is the stress
tensor; and \( f_i \) is the body force per unit mass.

The energy equation within the artery is given by
\[
(cp)_{bl} \frac{\partial T}{\partial t} + (cp)_{bl} \frac{\partial T}{\partial x_j} = \frac{\partial}{\partial x_j} \left( k_{bl} \frac{\partial T}{\partial x_j} \right) + q_s
\] (4)
where \( k_{bl} \) is the thermal conductivity (W/m/K) of blood.

Heat Source. A quasi-static spherical electrode is modeled as
an electromagnetical source for our coupled computational model.
The electrode has a diameter of 2.0 mm. To generate a scalable
heat source, a spherical ground electrode is placed 20.0 mm away
and the electric field generated in this condition [3,18]. The results
are parametrically fit to a Gaussian function using a Nelder–Mead
[15,19,20] optimization method of the form
\[
q_s(x,y) = A e^{-\left(x-x_m\right)^2+(y-y_m)^2/(a^2)}
\] (5)
where \( x, y, \) and \( z \) are the nodal coordinate values (cm), \( A \)
(=2.25 \times 10^6) is the peak value, whereas \( a \) (=11.6), \( b \) (=0.065),
and \( c (=1.17) \) the decide of the Gaussian heat source, \( x_m \)
 (=0) and \( y_m (=0) \) are the mean value of the \( x \) and \( y \) coordinates
(cm), respectively. \( A, a, b, c, x_m, \) and \( y_m \) were evaluated by the
Nelder–Mead scheme. The resulting heat source for a tissue vol-
ume of 0.52 \times 10^{-6} m^3 is shown in Fig. 3. The integrated value of
the Gaussian heat source is 14.1 \times 10^5 W/m^3 over a volume of
0.52 \times 10^{-6} m^3.

Material Properties. The present study has considered isotro-
pic tissue and blood properties at 37°C. The tissue and blood
properties [21] are density, \( \rho =1050 \) kg/m³; heat capacity, \( c_p \)
 =3600 J/kg/K; thermal conductivity, \( k =0.502 \) W/m/K; and
blood viscosity, \( \mu =0.0034 \) N s/m² [21].

Boundary Conditions. The thermal boundary condition at the
inlet of the artery is maintained at 37°C constant temperature. All
other surfaces of the computational domain have a zero heat-flux
condition. The initial temperature of the tissue and blood is also
considered to be 37°C.

Arterial velocities depend upon level of patient activity and
accordingly, for the present study, inlet arterial velocities of
1 cm/s and 13.8 cm/s, representing low flow and basal flow condi-
tions, are considered. These velocities are associated with non-
exercise conditions which are typically the case during the proce-
dure. A no-slip boundary condition is imposed along the artery
wall.

Hepatic perfusion rates in the literature are in the range of
0.2–1.1 mL/min/(mL of tissue) [22,23]. The larger values in-
clude contributions from the portal vein, and are useful for mod-
eling the convective effect of the portal vein, or the portal vein in
combination with the hepatic artery. A representative rate of
0.3 mL/min/(mL of tissue) is selected for characterizing tissue
perfusion. This hepatic perfusion rate is the total perfusion rate
minus the blood flow rates through the hepatic artery.

Finite Element Formulation. The equations governing fluid
flow and heat transfer are solved using the Galerkin finite ele-
ment (FE) method [24,25]. The partial differential equations and bound-
ary conditions are transformed to a system of algebraic equa-
tions that are solved to yield velocity, pressure, and temperature
\[
MV = K(U,T)V = F(U,T)
\] (6)
where \( K(U) \) is the global system matrix developed from the con-
servation of flow and energy; \( M \) is the mass matrix, and \( F \) is the
forcing function (including the body forces, e.g., sources, sinks,
etc.).

The steady-state velocity and pressure are solved first, and then
the transient temperature field is solved with the flow solutions.
The spatial integration is obtained using an iterative segregated
approach [24–27]. The continuous pressure projection form of the
segregated approach is used, employing Gauss–Seidel precondi-
tioning for symmetric and diagonal preconditioning for nonsym-
metric systems. Hybrid relaxation is used for the fluid flow (value
of 0.3 for all velocity components), and implicit relaxation
scheme was used for temperature calculations (value of 0).

The second-order streamline upwinding (SU) for all degrees of
freedom scheme developed by Hughes and Brooks is used for
stabilizing the solution. The SU scheme stabilizes the high-order
symmetric convection operators resulting from the Galerkin
method by explicitly adding numerical diffusion only along the
flow direction. This has the effect of weighting the convection
operators toward the upstream direction. The SU scheme reduces
the order of accuracy of the discretization to first order in the
streamwise direction but preserves the high accuracy of the Galer-
kin scheme in the cross-stream direction where no numerical dif-
fusion is added. Since convection typically dominates diffusion
along the flow direction, the added streamwise diffusion does not
adversely impact the accuracy of the solution in this direction.
As a consequence, the SU scheme only marginally degrades the order
of accuracy of the Galerkin scheme.

In solving for temperature, temporal terms are calculated with

Fig. 3 Gaussian-distributed RF heat source. The Gaussian
heat source in W/m² is plotted along the \( y \) axis.
an implicit time integration scheme (second-order trapezoid rule) in conjunction with an adaptive time-stepping methodology. The initial time step was 0.0001 and is controlled by a maximum rate of increase of 2.5%. The convergence criterion for velocity, temperature, and pressure is kept at $10^{-3}$.

Mesh independence is assessed by comparing the temperature distribution in the final working mesh, i.e., 500,000 element mesh, with values obtained using a refined mesh. After increasing the number of elements by 20%, the temperature distribution values for the maximum flow case changed by less than 1%. Computations are conducted on a Red Hat Linux (Version 7.3) workstation with dual Intel (Xeon IV) 2.4 GHz processors with 1.0 GB RAM. Complete analysis time for one set of the solutions is 7–8 h.

**Results**

Figure 4 shows temperature contours in the $x$-$z$ plane for different arterial inlet velocities when the tumor is at location-1 (Fig. 2(E)). It is evident from Fig. 4 that increasing the arterial velocity from 1 cm/s to 13.8 cm/s shrinks the volume of tumor heated considerably. In addition, the extent of the elevated temperature (red color) zone is significantly smaller for the higher arterial velocity. While the temperature contours are near symmetric in the $x$ direction (horizontal direction, Fig. 4), a noticeable asymmetry is present axially (vertical direction, Fig. 4).

The temperature profile along the $x$ direction is displayed in Fig. 5(A) for the tumor located very close to the bifurcation (location-1). Here, the temperature rise is plotted for the arterial inlet velocities of 0 cm/s, 1 cm/s, and 13.8 cm/s. In the case of no blood flow, the maximum temperature rise for a 20 min ablation time is $53.4^\circ$C (Fig. 5(A)). The profile width, defined as the distance over which the temperature rise drops to half the peak value, is approximately 5.65 mm. In the presence of blood flow, the maximum temperature rise is $47.7^\circ$C for lowest arterial velocity and $45.5^\circ$C for the highest. Taking the 20-min result to be steady state (little temperature change was observed beyond 20 min), the effect of convection of heat through the nearby arteries ($u = 13.8$ cm/s) is to reduce the steady-state peak temperature by $\sim 15\%$, where percentage reduction is defined as

$$\left(\frac{\Delta T_{\text{noflow}} - \Delta T_{\text{flow}}}{\Delta T_{\text{noflow}}} \times 100\right)$$

The corresponding profile widths vary from 5.7 mm for zero arterial velocity to 4 mm for the highest (13.8 cm/s). The temperature profile is symmetric along the $y$ direction as the tumor is located nearly equidistance from both the branches (Fig. 5(B)).

In contrast to the $y$-direction profiles, the temperature profile along the axial direction ($z$ direction) is highly asymmetric, as shown in Fig. 5(C). Temperature rise at the upper and lower boundaries of the tumor for the no flow case is 21°C. However, in the presence of blood flow, peak $\Delta T$ at the bottom of the tumor, close to the bifurcation (arrow # 1 in Fig. 5(C)), drops to 9°C ($\sim 57\%$ below the noflow value) for an arterial velocity of 1 cm/s. $\Delta T$ drops further to 6°C ($\sim 71\%$ below no flow) for the higher arterial velocity. However, near the upper tumor boundary (arrow # 2 in Fig. 5(C)), the temperature rise remains between 17°C and 20°C for higher and lower arterial flow rates, respectively.

Figures 6(A)–6(C) show the temperature profile along $x$, $y$, and $z$ direction for the tumor located 12 mm from the artery bifurcation region (location-2, Fig. 2(E)). Since the blood vessels are further away from the tumor, blood flow has minimum impact on the temperature rise in the tumor. For the arterial velocities of 1 cm/s and 13.8 cm/s, maximum temperature rise, in comparison with no blood flow condition, drop by $\sim 2.3\%$ and 3.8%, respectively (arrow # 1 in Fig. 6(A)). Similarly, negligible change in profile width is observed for both the flow rates.

Figure 7(A) shows the temperature distribution along the $x$ direction for tumor location-3 (Fig. 2(E)). In the absence of blood flow, maximum temperature rise is the same as calculated at the previous tumor locations, approximately 53°C. In the presence of blood flow, maximum temperature rise drops to 50°C for the lowest and 48°C for the highest arterial velocities, respectively. Blood flow ($u = 13.8$ cm/s) reduces the peak temperature rise by about 10% (arrow # 1 in Fig. 7(A)) and reduces the profile width by 33%, i.e., from 5.7 mm to 3.8 mm.

At this tumor location, an asymmetric temperature profile is obtained along the $x$ direction (Fig. 7(A)). The temperature rise at $x = -7$ mm, and $x = +7$ mm, representing the nearest and distant points on the tumor surface relative to the main artery (Fig. 7(A)), is 21°C for the no flow condition. For an arterial velocity of 1 cm/s, the temperature rise at the near side of the artery (arrow # 2 in Fig. 7(A)) drops to 11°C ($\sim 48\%$), while for the higher arterial velocity the temperature rise drops to 7°C ($\sim 67\%$). In contrast, the behavior along the $y$ direction shows insignificant asymmetry even in the presence of blood flow (Figs. 7(B)). However, temperature variation along the $z$ direction (Fig. 7(C)) is not perfectly symmetric due to the cooling effect of blood flowing through the right artery branch.

Figure 8 describes how arterial blood flow influences the rate of temperature rise during an ablation procedure. Temperature time history at the outer surface of the tumor located close to the bifurcation region (location-1, $z = -7$ mm) is shown in this figure. The temperature rise is seen to be independent of arterial flow during the first 40 s. In the absence of arterial flow, the temperature reaches 80% of its steady-state value (steady state $\sim 21^\circ$C) in about 600 s (arrow # 1 in Fig. 8). For a blood velocity of 1 cm/s, the 80% mark, relative to a steady-state value of 9°C, is reached in about 270 s (arrow # 2 in Fig. 8), and for the 13.8 cm/s blood velocity the 80% rise time (steady state $\sim 6.5^\circ$C) is around 160 s (arrow # 3 in Fig. 8).

In addition to steady-state temperature rise, tissue response is quantified in terms of a thermal dose and compared to accepted...
threshold values for cellular necrosis. From the transient temperature field, thermal dose values were calculated using an empirical method developed by Sapareto and Dewey [29]. In this method, an exponential relation exists between the tissue temperature and the exposure time that is required to coagulate the tumor cells. This relation is expressed quantitatively by the thermal dose parameter, which is expressed as

\[ t_{43}(x, y, z) = \int_{0}^{t_{\text{final}}} R^{43} - T(t) \, dt \]  

where \( t_{43} \) is the thermal dose at the reference temperature of 43°C, \( t_{\text{final}} \) is the treatment time, \( T(t) \) is the temperature field obtained from Eq. (1), and,

Fig. 5 Temperature rise within the tumor as a function of \( x \), \( y \), and \( z \) axis distances at time \( t =20 \) min for tumor location-1. \( z=0 \) denotes tumor center. Arterial velocities are 0 cm/s, 1 cm/s, and 13.8 cm/s. \#1=lower tumor boundary close to the bifurcation region; \#2=upper tumor boundary away from the bifurcation region.
\[
R = \begin{cases} 
0.5 & \text{if } T(t) \geq 43 \, ^\circ \text{C} \\
0.25 & \text{otherwise}
\end{cases}
\]

According to this relation, thermal dose resulting from heating the tissue to 43°C for 240 min is equivalent to that achieved by heating to 56°C for 1 s. Thermal dose threshold for complete cell damage is 240 min for a \(t_{43}\) of 43°C. This empirical relation is commonly used by researchers to quantify cell necrosis. However, this thermal dose calculation can be extended to evaluate cell damage using other criteria.

Thermal dose values at \(x=\pm 7 \, \text{mm}\) and \(z=\pm 7 \, \text{mm}\), for different arterial flow velocities, are plotted in Fig. 9 when the tumor is at location-1 (Fig. 2(E)). Since the temperature rise and the exposure time are exponentially related, arterial blood flow reduces the thermal dose value significantly. In the absence of blood flow, the thermal dose value at \(x=\pm 7 \, \text{mm}\) and \(z=\pm 7 \, \text{mm}\) is 3620 min, which indicates complete necrosis of the tumor volume. However, for an arterial velocity of 1 cm/s, the thermal dose value at \(z=-7 \, \text{mm}\) falls below the threshold level of 240 min. In addition,
at \( x = -7 \) mm thermal dose value is very close to the threshold level (Fig. 9). Similarly, for the higher flow rate, thermal dose values at \( z = -7 \) mm, \( z = +7 \) mm, and \( x = -7 \) mm fall below the threshold level. This implies that, due to the blood flow, there exists a partial volume of tumor which has not absorbed the required amount of energy to cause complete cell necrosis.

Although the tumor is located equidistant from the side branches in \( x-z \) plane (Fig. 2(A)), Fig. 9 shows that the thermal dose values obtained at \( x = -7 \) mm and \( x = +7 \) mm are different. This is due to the fact that the tumor is symmetric in the \( x-z \) plane, but not in the \( y-z \) plane (Fig. 2(B)), where the two vessel branches are not in the same plane. As a result, one of the branches has more influence on the ablation procedure than the other one. This can also be seen in Fig. 5(A) where there is approximately a 2°C difference in temperature rise at \( x = -7 \) mm and \( x = +7 \) mm. Since thermal dose is exponentially related to the time history of temperature, this small difference gets amplified in the thermal dose calculations (Fig. 9) [30–33].

**Discussion**

The significant asymmetries across \( z = 0 \) in the temperature (Fig. 5(c)) and thermal dose (Fig. 9) profiles demonstrate the impor-
tance of direct physical modeling through large vessels, since a
pure lumped-parameter approach would show the same thermal
effects on the top and bottom of the heated tumor. While the
thermal dose is adequate (>240 min) to achieve complete cell
destruction on the top of the tumor (z=+7 mm), it is below
threshold on the bottom of the tumor (Fig. 9) in the presence of
arterial flow, potentially leading to tumor recursion.

Incorporating the specific vascular geometry in the calculation
is likewise critical, since locations 1 and 3 are approximately the
same distance from the large vessel, but the closer spacing of the
curves for different flow rates in Fig. 7(a) relative to the spacing
in Fig. 5(a) indicates a weaker dependence upon arterial blood
flow. This is due partly to the larger surface area available for heat
transfer at the bottom of tumor location 1 (top of the bifurcation),
as well as the presence of the daughter vessels at the top part of
tumor location 1.

If the blood flow through the large vessel is defined to be in-
consequential wherever the temperature rises for the three blood
velocities: 0 cm/s, 1 cm/s, and 13.8 cm/s are within 1% of one
another, it can be concluded that blood flow within the bifurcating
artery ceases to affect the ablation procedure beyond a distance of
about 15 mm above the bifurcation. This value is derived by ob-

Fig. 8 Temperature time history along the outer surface of the tumor
located near the bifurcation point (location-1). Arterial velocities are
0 cm/s, 1 cm/s, and 13.8 cm/s. #1, #2, #3=time at which the temperature
rise attains 80% of the steady state value.

Fig. 9 Thermal dose values as a function of artery inlet velocities (1 cm/s
and 13.8 cm/s) at z=±7 mm and x=±7 mm
serving the axial dependence of the temperature profile (Fig. 6(c)) for location 2, and noting that the three curves converge at a distance of about 3 mm above the center of the heated volume (arrow # 1 in Fig. 6(c)), or about 15 mm above the surface of the arterial bifurcation. The 15 mm value assumes a translation purely along the axial direction; additional movement toward one of the daughter vessels (x, y translation) could render the large-scale blood flow once again influential.

For the case of the tumor located to the side of the parent vessel (location 3), the distance where the arterial flow becomes significant is around 18 mm (arrow # 3 in Fig. 7(A)) from the vessel. Since the energy equation (Eq. (1)) is linear in the amount of absorbed power, these estimates for the region of influence of arterial flow apply regardless of the applied electrical power. Whether cell necrosis is achieved will of course be a function of the power level.

Limitations of the present model include the use of a simplified, spherically symmetric heat source. Future generations of the model will include a more realistic electrical-field configuration, such as that due to a four-tine probe considered by Tungjitkusolmun et al. [5]. Also, Chang [18] has demonstrated that the temperature dependence of the electrical conductivity can be significant, and should be addressed in the model. The electric and thermal fields become coupled, and the electric field equation must be solved simultaneously with those of fluid and heat transfer.

The present model also does not account for bulk perfusion changes due to tissue necrosis, nor does it directly model vessels smaller than the bifurcating artery. Hence, the model is most useful for providing insight into if ablation when convective heat transfer is dominated by the effects due to a single vessel. The reconstruction technique is applicable to more general situations where additional vessels contribute, as in the case of a tumor located at comparable distances to multiple vessels. “Countercurrent’’ heat transfer [8] between a vein–artery pair would be automatically included in direct physical modeling, provided all relevant vessels could be resolved in the medical image. A price paid for the very general (or accurate) simulation capability is an increase in computation time for each vessel directly modeled.

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Nomenclature

\( \begin{align*} X, Y, Z &= \text{nodal coordinate values, cm} \\
A, b, c &= \text{constants of Gaussian approximation of heat source} \\
x_m &= \text{mean value of } x \text{ coordinates, cm} \\
y_m &= \text{mean value of } y \text{ coordinates, cm} \\
P &= \text{pressure, N/m}^2 \\
c_p &= \text{tissue specific heat, J/kg/K} \\
u &= \text{nodal velocity component, cm/s} \\
k &= \text{thermal conductivity of tissue, W/mK} \\
\alpha &= \text{thermal diffusivity of tissue, m}^2/\text{s} \\
\rho &= \text{density of tissue, kg/m}^3 \\
s &= \text{stress tensor, N/m}^2 \\
F &= \text{body force per unit mass, N/kg} \\
K &= \text{global system matrix developed from the conservative of flow and energy} \\
M &= \text{mass matrix} \\
\omega &= \text{blood perfusion, ml/min/ml} \\
T_m &= \text{blood temperature, } ^\circ \text{C} \\
T_i &= \text{initial temperature, } ^\circ \text{C} \\
c_{bl} &= \text{blood heat capacity, J/kg/K} \\
\rho_{bl} &= \text{blood density kg/m}^3 \\
k_{bl} &= \text{thermal conductivity of blood, W/mK} \\
\mu &= \text{viscosity of blood, N s/m}^2 \\
q_s &= \text{Gaussian heat source, W/m}^3 \end{align*} \)

Appendix

In this section, an attempt has been made to come up with an alternative model to the Pennes bio-heat equation. In this approach the hepatic tissue is modeled as a porous region. Cooling due to perfusion is treated using direct physical modeling of the heat and momentum transfer within vessel and tissue domains.

Governing Equations

The hepatic tissue is modeled as a porous medium when the tissue perfusion is treated with a statistical approach. Within the porous medium, fluid and solid regions are identified. If \( V_f \) represents the volume of fluid and \( V_t \) represents the solid volume, such that the total volume, \( V_t \) is given by

\[ V_f = V_f + V_t \]

Porosity is defined by

\[ \phi = \frac{V_f}{V_t} \]

In order to analyze the porous tissue region, the volume elements are considered sufficiently large compared with the length scale of an individual pore. With these assumptions and steady-state conditions, the conservation of mass is expressed by the modified continuity relation

\[ \nabla \cdot \bar{u} + \frac{\dot{q}}{\rho} = 0 \]

\( \bar{u} \) is the average of the fluid velocity (m/s) over \( V_f \) and \( q \) is the mass source term (kg/s/(ml of tissue)) accounting for the blood perfusing through the tissue volume. The \( q \) value is calculated based on representative tissue perfusion of 0.3 mL/min/(ml of tissue). The momentum equation is expressed by Darcy’s law

\[ \bar{u} = -k \nabla p \]

where \( k \) is the hydraulic conductivity tensor and \( k_i = k \mu \) is the permeability (m²); and \( \mu \) is the dynamic viscosity of fluid (N s/m²). \( k \) is dependent on the pore geometry and interstitial matrix of the porous medium and is a measure of the conductance of the material to fluid flow.

The following energy equation for porous media is also used to determine the temperature distribution in the tissue domain

\[ (pc_p) \frac{\partial T}{\partial t} + (pc_p)u_j \frac{\partial T}{\partial x_j} = \frac{\partial}{\partial x_j} \left( k_i \frac{\partial T}{\partial x_j} \right) + q_s \]

The effective properties are related to fluid and solid matrix properties by the relations

\[ (pc_p)_i = \phi pc_p + (1 - \phi)(pc_p)_s \]

\[ (k_i) = \phi k + (1 - \phi)(k)_s \]

where \( \phi \) is the porosity, properties without subscripts are those of the blood, and properties with the subscript \( s \) are of the tissue solid.
The permeability was selected so that the tissue perfusion rate, defined as the net rate of blood flow out of the tissue volume divided by the total tissue volume, was in the range of published perfusion rates. A permeability of $10^{-10}$ m$^2$ was required to achieve the defined level of perfusion. The authors are unaware of published values for the porosity of liver specifically. Thus, a typical tissue porosity of 0.2 is selected.

**Boundary Conditions**

At the side (radial) boundaries of the computational domain, a condition of zero normal stress was imposed. As the computational volume is in the form of short cylinder, a symmetry condition is imposed between the upper and lower tissue boundaries. This results in zero perfusion out of the top and bottom of the tissue volume. All percolating blood exits the tissue volume radially outward. Blood crosses the top and bottom of the computational cylinder only through the bifurcating artery. A no-slip boundary condition was imposed along the artery wall.

Governing equations and the boundary conditions for the vessel domain is similar to the one used for the lumped parameter approach.

**Results**

Figure 10 shows temperature plots along $y$ and $z$ directions for three different conditions: (a) no convection and scalar perfusion, (b) $u=13.8$ cm/s as inlet velocity including vector perfusion and (c) $u=13.8$ cm/s as inlet velocity including vector perfusion.
lature including perfusion as a scalar quantity (also shown in Figs. 6(A) and 6(B)); (b) \( \dot{u} = 13.8 \text{ cm}^3/\text{s} \) as inlet velocity including perfusion as a scalar quantity (lumped parameter approach) shown in Figs. 6(A) and 6(B); and, (c) \( \dot{u} = 13.8 \text{ cm}^3/\text{s} \) as inlet velocity including the directionality of the perfusion (statistical approach). In the case of no blood flow (case (a)), the maximum temperature rise for a 20-min ablation time is 53.4°C (Fig. 10(a)). In comparison to the no-flow case, the effect of scalar perfusion (case (b)) is to reduce the peak temperature of the tumor to 51°C. If the directionality of the perfusion (case (c)) is accounted for the peak \( \Delta T \) drops further to 49°C (8.2% drop in comparison to the no-flow case). The difference in temperature rise calculated from scalar and vector models is within \( \sim 4\% \), when the cooling effect due to the blood flow through the large artery is significant. \( \Delta T \) rise in the \( z \) direction shows similar variation for all the three models. In the region of ablation near the large vessels having significant convection, the vector approach has insignificant variation when compared with the scalar approach. Considering this, the vector approach that requires increased computation time has not been pursued further in this study.

References