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FINITE ELEMENT ANALYSIS OF RADIO-FREQUENCY ABLATION IN A RECONSTRUCTED REALISTIC HEPATIC GEOMETRY

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ABSTRACT

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 to heat energy and thus, a rise in temperature, with the goal of eventual tumor necrosis. Potential complications from the procedure include insufficient heating of large tumors, resulting in tumor recursion, as well as excessive thermal damage to healthy tissue. Mathematical models are valuable in predicting the temperature rise within the organ during RF ablation, thereby enhancing the success rate of the procedure. Eventually, models can be used to guide ablation procedures, by predicting the optimal set of operational parameters e.g., catheter probe geometry and placement, given patient-specific information. The present study focuses on the analysis of temperature rise within a reconstructed model of a realistic three-dimensional (3D) section of a porcine liver during RF ablation.

This study calculates the effect of blood flow through arteries as well as perfusion through the liver on the time-dependent temperature distribution near the RF ablation probe (Figure 1). For a time duration of 30 min of an ablation procedure, a temperature of about 80°C could be achieved over a diameter of about 4 cm with the present RF probe. As an initial step, the present study includes isotropic hepatic tissue and blood properties.

INTRODUCTION

The term "radiofrequency ablation (RF) probe" refers to a class of medical devices operating between 460-550 kHz that delivers therapeutic energy into soft tissues.

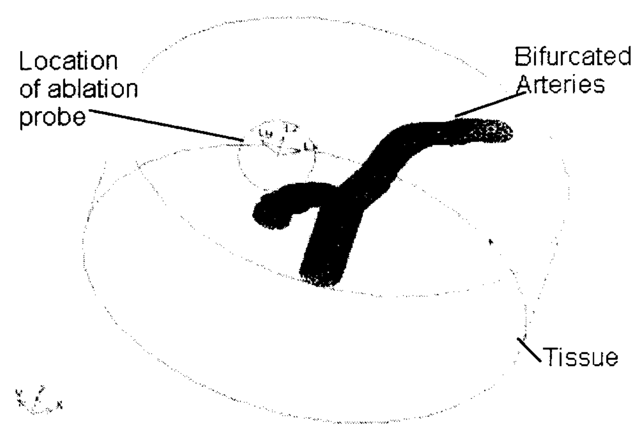

Figure 1: Reconstructed bifurcated arteries with surrounding hepatic tissues. The RF ablation probe is inserted in the spherical tissue domain, considered to be the ablation zone.

Typically, energy is injected through the probe, into soft tissues, to a ground electrode that is usually placed on the surface of a patient, thereby completing the electric circuit. Although radiofrequency ablation has been widely used for several decades, significant advancement in understanding the physics has only recently been made.

In computing heat transfer during ablation procedures, the effect of blood perfusion is indirectly modeled as a source term using a lumped heat-transfer coefficient in the energy equation. This coefficient is assumed to apply throughout the entire liver [1, 2]. While such an approach is sufficient for heat transfer through an ensemble of capillaries, it is unlikely to be accurate in the case of ablation near a large blood vessel. In this study, both convective blood flow and heat transfer within the conjugate domains of large vessels and surrounding porous hepatic tissues are solved directly using the coupled mass, momentum, and energy transport equations, thus, improving on the previous studies. This direct-simulation approach requires information regarding the vascular geometry of the treated organ. MRI scans of the affected tissue are acquired, followed by segmentation and generation of a computational mesh. The technique will be illustrated below for the case of an excised porcine liver model.

Accurate analysis of RF ablation procedures also requires a realistic representation of the electric field generated by the RF catheter. In principle, this field is coupled to the temperature and fluid-flow fields, as the electric conductivity is temperature dependent. For the present study, as an initial step, electric field values which have been previously determined are equated to a distributed heat source either as a constant or, more accurately, as a gaussian function over a tissue domain. The temperature rise over time induced by the heat source near a large blood vessel within the porcine liver is calculated, and the effect of blood perfusion on the temperature field is assessed.

NOMENCLATURE

c_p	Specific heat, J/kg.K
k_i	Permeability, m^2
k_e	Thermal conductivity, W/m.K
p	Fluid pore pressure, N/m^2
q_s	Heat source from ablation probe, W/m^3
T	Temperature, ^0C
u	Velocity, m/s
u	Average velocity, m/s
V	Volume, m^3
μ	Dynamic viscosity, Ns/m^2
?	Density, kg/m^3
φ	Porosity

subscripts:

e	Effective property of both the fluid and solid
f	Fluid properties
i, j, k	x, y, & z direction for a 3 – D problem
s	Solid matrix
t	Total

METHODOLOGY

The analysis of the RF ablation procedure is accomplished through a Finite-Element (FE) solution to the fluid-flow and energy equations within the reconstructed liver geometry having blood flow through the arteries and blood perfusion through the porous hepatic tissue. This section describes the methodologies for acquiring the physiologic geometry and for numerically representing the transport processes.

Image Reconstruction and Meshing: Using a small bore MRI, a total of 128 images spaced 625 microns apart are obtained from an excised porcine liver segment. In order to identify the geometry of the arteries and surrounding tissue, images are segmented by automatic upper and lower threshold values [3], which are based on image contrast. All gray colored pixels between the threshold values are treated as hepatic tissue, thus identifying the artery supplying blood to the liver. 3D reconstruction of the geometry is achieved by a region growing technique from the segmented MR scans. After checking the smoothness of the 3D 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 containing real vertices, edges, and surfaces are imported for automatic, linear, tetrahedral mesh generation [4]. The excised liver section is of a cylindrical shape (Figure 1) with a height of 72 mm and diameter of 180 mm, containing about 190,000 elements.

Equations: In the present study the hepatic tissue is modeled as a porous medium with blood perfusion. Within porous materials, there are two distinct regions of interest: fluid and solid. If V_f represents the volume of fluid and V_s represent the solid volume, such that the total volume, V_t is given by,

$$V_t = V_f + V_s. \quad (1)$$

Porosity is defined by,

$$\phi = V_f / V_t \quad (2)$$

In order to construct a continuum model, 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 continuity relation,

$$\nabla \cdot \bar{u} = 0 \quad (3)$$

where ∇□is the divergence operator and $\bar{u}$ is the average of the fluid velocity over V_t. The momentum equation is expressed by Darcy's law,

$$\bar{u} = -k \nabla p \quad (4)$$

where k is the hydraulic conductivity tensor, $k_i = k\mu$ is the permeability, and ∇ is the gradient operator. 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. An implicit assumption is that the ambient temperature does not change the fluid density, and subsequently the fluid flow profile.

The following energy equation for porous media is also used for temperature distribution analysis:

$$(\rho c_p)_e \frac{\partial T}{\partial t} + \rho c_p u_j \frac{\partial T}{\partial x_j} = \frac{\partial}{\partial x_j}\left(k_e \frac{\partial T}{\partial x_j} \right) + q_s \qquad (5)$$

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

$$(\rho c_p)_e = \phi \rho c_p + (1-\phi)(\rho c_p)_s \qquad (6)$$

$$k_e = \phi k + (1-\phi) k_s$$

(7)

Where properties without subscripts are those of the fluid. First, the momentum equation is solved to calculate the velocity field in the liver tissue. Subsequently, the velocity field is used as input and coupled with the energy equation to get the temperature field in the liver domain at different time. Within the arterial region, the mass, momentum and energy equations of fluid are applied. The governing equations are solved [5] with a segregated solver and a second order trapezoidal time integration scheme.

Model Parameters: In the present study the RF probe is modeled as a spherical heat source with an average value of 2.7×10^5 W/m^3 over a volume of 14.1×10^{-6} m^3. The boundary condition imposed on all external surfaces is zero heat flux except at the inlet of the artery and the tissue near it, where a constant temperature of 37°C is used. The average blood perfusion rate through the hepatic tissue is maintained at 0.6 ml/min/ml volume of liver [1,2], which, in turn, resulted in a blood velocity of 0.124 m/s is for the reconstructed inlet diameter of the artery. A stress free boundary condition is used at all other boundaries.

Tissue and Blood Properties: The present study considered isotropic tissue and blood properties at 37°C. The tissue and blood properties [6] are: density, ? = 1050 kg/m^3; capacity, c_p=3600 J/kg.K; conductivity, k = 0.502 W/m.K. The infinite shear-rate blood viscosity, μ=0.0034 Ns/m^2. As an initial estimate the hepatic tissue porosity is considered to be 0.2 and the isotropic permeability is 10^{-10} m^2.

RESULTS AND DISCUSSION

Figure 2 represents the temperature variation in x-direction with time. As expected, there is a rapid increase in temperature during initial times, e.g., 8°C increase in first 2 min whereas only 5°C increase occurs between 30 min and 60 min. For a time duration of 30 min. of ablation procedure, a temperature of about 80°C could be achieved over a diameter of about 4 cm with the present RF probe.

The effects of blood flow rate variation, different probe design, and direct coupling of electrical field with transport equations are presently under evaluation. Anisotropic hepatic tissue properties including permeability and temperature dependent thermal conductivities will also be evaluated. The impact of more detailed reconstructed geometry including hepatic arteries, portal veins, hepatic veins, hepatic sinuses surrounded the tissue will be included in future studies.

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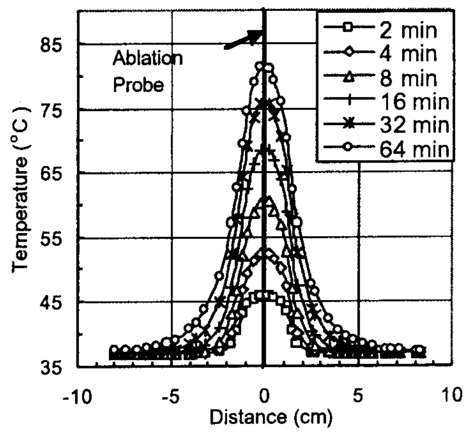

Figure 2: Temporal variation of temperature distribution in x-direction

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