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A Two-Scale Model of Radio-Frequency Electrosurgical Tissue Ablation

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Abstract

Radio-frequency electrosurgical procedures are widely used to simultaneously dissect and coagulate tissue. Experiments suggest that evaporation of cellular and intra-cellular water plays a significant role in the evolution of the temperature field at the tissue level, which is not adequately captured in a single scale energy balance equation. Here, we propose a two-scale model to study the effects of microscale phase change and heat dissipation in response to radiofrequency heating on the tissue level in electrosurgical ablation procedures. At the microscale, the conservation of mass along with thermodynamic and mechanical equilibrium is applied to obtain an equation-of-state (EOS) relating vapor mass fraction to temperature and pressure. The evaporation losses are incorporated in the macro-level energy conservation and results are validated with mean experimental temperature distributions measured from electrosurgical ablation testing on *ex vivo* porcine liver at different power settings of the electrosurgical instrument. Model prediction of water loss and its effect on the temperature along with the effect of the mechanical properties on results are evaluated and discussed.

Keywords

Electrosurgery; two-scale model; evaporation model; soft tissue ablation

1. Introduction

Alternating current in the radiofrequency range of 300 kHz – 1 MHz is used in electrosurgical procedures to generate tissue heating and conduct various surgical procedures [1]. These procedures, such as ablation of metastatic liver tumors [2] and resection of organs [3] can involve peak temperatures greater than 100 °C. Understanding the mechanisms of heat and mass transfer in biological tissue during electrosurgery are of critical importance in developing models that can predict and simulate tissue damage accurately. Computational models are often used to assess the safety of electrosurgical

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procedures and to develop new instrument designs [5], but the results often have significant differences from experimental data.

The challenge in accurately predicting temperature changes arises from the need to model energy dissipation that is significantly affected by changes in temperature dependent tissue properties, latent heat loss, and phase change [6][7]–[8]. Previous work measuring the apparent specific heat of *ex vivo* porcine liver based on a heat conduction model found that energy losses due to latent heat and phase change become dominant at temperatures above 80°C [9]. Yet, existing models do not account for these phenomena in a rigorous manner, as discussed in the next paragraph. One main challenge is the fact that tissue heating during electrosurgery is volumetric resulting from electrical conductivity losses within the tissue where the process is inherently multiscale in nature. Hydrated soft tissues contain more than 70% water by mass [10] which can be considered to be distributed in small pores on the order of 10^{-7} m within a solid matrix [11]. Ohmic heating leads to evaporation of intracellular and cellular water which plays a significant role in the temporal evolution of the temperature. Additionally, the electric field on which this heat generation depends, falls off as the inverse of the square of the distance from the electrode. This creates a high power density in the vicinity of the electrode with sharp gradients of temperature and pressure and large evaporation losses [4]. Another challenge is related to the change of phase of water. When heated through an ablation process, phase change of liquid water to vapor can occur through equilibrium vaporization which takes place at a liquid-vapor interface, with liquid water becoming vapor at the saturation temperature and pressure. Normal boiling requires the presence of heterogeneities that provide nucleation sites that allow the growth of vapor bubbles, however, in tissue the mobility of vapor bubbles is inhibited by the extracellular matrix, thus boiling is confined [12]. Thus, the energy required to drive bubble growth must be large enough to overcome both surface tension and the elastic strain energy required to deform the solid matrix surrounding the bubble [13].

A few studies in the literature have considered the effect of evaporation on energy dissipation when simulating energy based surgical procedures [6], [14]–[18] such as electrosurgery or laser ablation. Abraham *et al.* considered evaporation through a piece-wise description of the tissue enthalpy where in the 99 – 100 °C range, the fraction of water evaporated increases linearly with temperature until all the water in the tissue has evaporated at 100 °C [14]. In other studies, such as the one's by Yang *et al.* the effect of energy losses due to latent losses are included in a modified apparent specific heat [16]. This was based on measuring the water content of liver tissue after an ablation process. The variation in water content was then correlated to temperature and fitted to an approximate numerical model [17]. Chen *et al.* used Yang's effective specific heat in modeling vessel sealing in bipolar electrosurgery at high temperatures [3]. A mechanistic model proposed by Majaron *et al.* modeled the phase change as confined boiling where water is assumed to be trapped in spherical cavities within an elastic medium. As the tissue is heated, vaporization causes the pressure within those cavities to increase creating a mechanical equilibrium with the surrounding solid [18]. The one dimensional model produced more realistic prediction range of the temperature evolution for 1 ms laser heating of skin tissue, where temperatures reached 200°C in 300 μ s, but the model was not validated against experimental results.

Electrosurgical heating has less power density than laser heating and time needed for temperatures rise to 100°C is on the order of seconds.

In this paper, a two-scale model has been proposed to study the effects of microscale level evaporation and heat dissipation on the thermal response of soft tissue to radiofrequency heating in electrosurgical procedures. Soft tissue is assumed to be macroscopically homogeneous, but a microscopically heterogeneous body consisting of fluid trapped in microscopic pores. At the microscale level, the conservation of mass along with the thermodynamic and mechanical equilibrium is applied to obtain an equation-of-state (EOS) relating vapor mass fraction to temperature and pressure. The vapor mass fraction is then obtained by solving local microscopic energy balance equation along with EOS. The temporal evolution of the temperature field at macroscale is obtained using effective parameters that accounts for the evaporation losses based on microscale calculations.

The paper is organized as follows. The two-scale model of radiofrequency heating of soft tissue along with the micromechanical model is presented in Section 2. The results are validated with mean experimental temperature distributions measured from electrosurgical ablation testing on *ex vivo* porcine liver at different power settings in Section 3, followed by conclusions and future scope of work in Section 4.

2. Mathematical model

In the proposed two-scale model, evaporation is described through an equation of state obtained from the thermodynamic and mechanical equilibrium of the solid, water and vapor phases within microscopic pores in the tissue. The evaporation losses are incorporated in the microscale energy balance to obtain equilibrium thermodynamic states of the micro pore system. The temperature distribution at the macroscale tissue level is then obtained by solving the transient heat equation using effective thermomechanical properties and effective heat generation provided by the microscale model.

In the following sections, we first present the macroscale model of the conservation of electric charge and heat conduction in the tissue. Next, we describe the micromechanical model, which is based on Majaron et al.'s [18] work on the single scale modeling of laser ablation of soft tissue, followed by computation of effective parameters.

2.1. Macroscale model

At the macroscale we consider an open bounded domain $\Omega \in \mathbb{R}^3$ with external boundary Γ . The electric potential and temperature fields must satisfy the following local forms of the balance laws for all $\mathbf{x} \in \Omega$ and time $t \in (0, \tau)$.

Electric charge conservation—The equivalent root mean square of the voltage and current are considered. Conservation of electrical charge provides the equation for the potential distribution V in the tissue. The boundary value problem for charge conservation is

$$\text{Find } V: \bar{\Omega} \rightarrow \mathbb{R} \text{ such that}$$

$$\begin{aligned}\nabla \cdot \mathbf{J} &= 0 \text{ in } \Omega \\ \mathbf{n} \cdot \mathbf{J} &= j_n \text{ on } \Gamma_j\end{aligned}\quad (2.1)$$

$$V = \bar{V} \text{ on } \Gamma_V$$

with boundary $\Gamma = \Gamma_j \cup \Gamma_V$ and $\Gamma_j \cap \Gamma_V = \emptyset$, and $\bar{\Omega} = \Omega \cup \Gamma$, where $\mathbf{J}$ is the current density, j_n is the current density normal to the Neumann boundary Γ_j , $\mathbf{n}$ is the unit outward normal to Γ_j , and $\bar{V}$ is the prescribed electric potential on the Dirichlet boundary Γ_V .

At the frequency used in electrosurgical procedures, capacitive losses are negligible and heat generation can be considered to be mainly resistive [6]. The current density can then be described by Ohm's law as,

$$\mathbf{J} = \sigma \mathbf{E}, \quad (2.2)$$

where σ is the temperature dependent electrical conductivity, and $\mathbf{E}$ is the electric field given as,

$$\mathbf{E} = -\nabla V. \quad (2.3)$$

The volumetric heat generated due to radiofrequency electrical excitation $\dot{q}_e$, required to solve the microscale model, is

$$\dot{q}_e = \mathbf{J} \cdot \mathbf{E}. \quad (2.4)$$

Energy balance—The temperature distribution in the tissue is obtained by solving the transient heat conduction equation. The initial boundary value problem may be stated as,

Find $T: \bar{\Omega} \times [0, T] \rightarrow \mathbb{R}$ such that

$$\begin{aligned}\rho_{eff} c_{eff} \frac{\partial T}{\partial t} + \nabla \cdot \mathbf{q} &= \dot{q}_{eff} \text{ in } \Omega \times]0, \tau[\\ -\mathbf{q} \cdot \mathbf{n} &= q_n \text{ on } \Gamma_q \times]0, \tau[\\ T &= \bar{T} \text{ on } \Gamma_T \times]0, \tau[\\ T(\mathbf{x}, 0) &= T_0(\mathbf{x}) \quad \mathbf{x} \in \Omega\end{aligned}$$

$$(2.5)$$

with boundary $\Gamma = \Gamma_q \cup \Gamma_T$ and $\Gamma_q \cap \Gamma_T = \emptyset$, where $\mathbf{q}$ is the heat flux vector, $(\dot{q}_{eff})$ is the rate effective heat input per unit volume determined from the microscale model described next, q_n is the component of heat flux normal to the Neumann boundary Γ_q , $\mathbf{n}$ is the unit normal to Γ_q , $\bar{T}$ is the prescribed temperature at the Dirichlet boundary Γ_T , T_0 is the initial temperature, and ρ_{eff} and c_{eff} are the effective density and effective specific heat capacity of the tissue, respectively. The heat flux is defined in terms of the temperature gradient using generalized Fourier's law as

$$\mathbf{q} = -k_{eff} \nabla T, \quad (2.6)$$

where k_{eff} is the effective thermal conductivity of the tissue.

The effective parameters (ρ_{eff} , c_{eff} , k_{eff}) along with effective heat generation $(\dot{q}_{eff})$ are obtained by solving the cellular level microscale model presented in the next section.

2.2. Microscale model

Associated with each macroscale spatial point $\mathbf{x} \in \Omega$ of the tissue is a microscopic representative volume element (RVE), where fluid is assumed to be trapped in a non-absorbing elastic medium (Fig. 1). Initially, the pore of volume V_0 is filled with water at ambient temperature T_a and atmospheric pressure p_0 . With Joule heating, the temperature of the water increases until the boiling temperature T_0 (at pore pressure P_0) is reached, beyond which a mass fraction x of the water converts into vapor leading to expansion of the pore under added pressure. The water vapor is assumed to behave as an ideal gas. For an ideal gas approximation of the vapor within the pore, the equilibrium vapor pressure P is related to the temperature T through the Clausius-Clapeyron equation [19] given by,

$$P = P_0 \exp \left[-\frac{\lambda}{R_w} \left(\frac{1}{T} - \frac{1}{T_0} \right) \right], \quad (2.7)$$

where λ is the average specific enthalpy of evaporation in the interval $[T_0, T]$.

The current volume V of the pore is the volume of the liquid water (V_w) and vapor (V_v) and can be related to the initial pore volume V_0 by applying mass conservation of the water and vapor $\rho_w V_0 = \rho_w V_w + \rho_v V_v$, where ρ_v is the density of water vapor obtained through ideal gas law. By considering the mass fraction of liquid water evaporated $x = \rho_v V_v / \rho_w V_0$, the relation between current and initial volume can be given by,

$$V = V_0 \left[\left(\frac{R_w T \rho_w}{P} - 1 \right) x + 1 \right], \quad (2.8)$$

where R_w is the specific gas constant of water vapor, ρ_w is the density of water at pressure P and temperature T .

By applying mechanical equilibrium for a single spherical cavity in an infinite elastic medium [20], the current volume of the pore can also be approximated as,

$$V = V_0 \left[1 + \frac{1+\nu}{2E}(P - P_0) \right]^3, \quad (2.9)$$

where E is the young's modulus of the tissue and ν is the Poisson's ratio.

Equation-of-state—The two relations for the relative volume change of the pore volume in Eqs. (2.8) and (2.9) lead an equation-of-state (EOS) that relates the pore pressure P and temperature T with the vapor content x as,

$$x = \left[\left(1 + \frac{1+\nu}{2E}(P - P_0) \right)^3 - 1 \right] \left(\frac{R_w T \rho_w}{P} - 1 \right)^{-1}, \quad (2.10)$$

Energy balance—The energy balance is found by applying the First Law of Thermodynamics to a closed system, assuming a quasi-state expansion of the micro pore in the RVE. The change of specific internal energy over time interval dt is expressed in terms of the total energy applied to the system and the work done by the pore on the surrounding tissue where $du = dq_e - pdv$. This can be written in terms of the specific enthalpy where $dh = dq_e + vdP$. In incremental form, the energy balance is then given by,

$$m[\lambda \Delta x + x(c_v + R_w)\Delta T + (1-x)c_w \Delta T] - \frac{1}{\rho_{eff}} \Delta P + (1-m)c_s \Delta T = \frac{1}{\rho_{eff}} \dot{q}_e \Delta t, \quad (2.11)$$

where m is the total water and vapor mass fraction, and c_s , c_w , and c_v represent the specific heat capacity of the solid, water, and vapor phases respectively. The terms x and P are dependent only on temperature per Eqs (2.7) and (2.10), they can be expressed as $x = (dx/dT) T$ and $P = (dP/dT) T$.

Effective quantities—The effective density ρ_{eff} , specific heat capacity c_{eff} , and thermal conductivity k_{eff} of the tissue at $\mathbf{x} \in \Omega$ are related to the vapor fraction x by,

$$\rho_{eff} = mx\rho_v + m(1-x)\rho_w + (1-m)\rho_s, \quad (2.12)$$

$$c_{eff} = mx c_v + m(1-x)c_w + (1-m)c_s, \text{ and } \quad (2.13)$$

$$k_{eff} = mxk_v + m(1-x)k_w + (1-m)k_s, \quad (2.14)$$

where the subscripts v , w , and s represent properties corresponding to the vapor, water, and solid phase, respectively.

Effective heat generation—The effective heat input at a macroscopic spatial point $\mathbf{x} \in \Omega$ is computed by

$$\dot{q}_{eff}(\mathbf{x}, t) = \rho_{eff} c_{eff} \dot{T}. \quad (2.15)$$

where $\dot{T}$ is rate of temperature change obtained from the microscale solution.

2.3. Discretized equations and numerical solution approach

The following describes the weak form of the electric charge conservation and temperature evolution equations followed by the numerical solution approach.

Macroscale finite element formulation—Let

$\mathcal{V}_v = \{w(\mathbf{x}) | \mathbf{x} \in \bar{\Omega}, w(\mathbf{x}) \in H^1(\Omega), w(\mathbf{x}) = 0 \text{ on } \Gamma_V\}$, be the space of weighing function, where H^1 is the first order Hilbert space. Let $\mathcal{S}_v = \{V(\mathbf{x}) | \mathbf{x} \in \bar{\Omega}, V(\mathbf{x}) \in H^1(\Omega), V(\mathbf{x}) = \bar{V}(\mathbf{x}) \text{ on } \Gamma_V\}$ be the space of trial solution. The weak form of the electric charge conservation law in equation (2.1) is given as

$$\text{Find } \{V(t) \in \mathcal{S}_v\}, \text{ such that for all } w \in \mathcal{V}_v$$

$$\int_{\Omega} \nabla w \cdot \mathbf{J} d\Omega = \int_{\Gamma_j} w j_n d\Gamma. \quad (2.16)$$

The boundary conditions are given in Eq. (2.1).

Further, let $\mathcal{V}_t = \{w(\mathbf{x}) | \mathbf{x} \in \bar{\Omega}, w(\mathbf{x}) \in H^1(\Omega), w(\mathbf{x}) = 0 \text{ on } \Gamma_T\}$ be the space of the weighing function and $\mathcal{S}_t = \{T(\mathbf{x}, t) | \mathbf{x} \in \bar{\Omega}, T(\mathbf{x}, t) \in H^1(\Omega), T(\mathbf{x}, t) = \bar{T}(\mathbf{x}, t) \text{ on } \Gamma_T\}$ be the space of trial solution. The weak form of the heat equation in (2.5) may be stated as [21],

$$\text{Find } T(t) \in \mathcal{S}_t, t \in [0, \tau] \text{ such that for all } w \in \mathcal{V}_t$$

$$\int_{\Omega} w \rho_{eff} c_{eff} \frac{\partial T}{\partial t} d\Omega + \int_{\Omega} \nabla w \cdot \mathbf{q} d\Omega = \int_{\Omega} w \dot{q}_{eff} d\Omega + \int_{\Gamma_q} w q_n d\Gamma \quad (2.17)$$

The initial and boundary conditions are given in Eq. (2.5).

We assume finite element discretization of the weighting function, electric potential, and temperature fields as

$$w(\mathbf{x}) = \sum_I N_I(\mathbf{x}) w_I, V(\mathbf{x}) = \sum_I N_I(\mathbf{x}) V_I, \text{ and } T(\mathbf{x}, t) = \sum_I N_I(\mathbf{x}) T_I(t), \quad (2.18)$$

respectively, where $N_I(x)$ is the shape function of node I , V_I is the nodal electric potential, T_I is the nodal temperature, and w_I is a constant.

The discretized weak form of the charge conservation law (Eq. (2.16)) is given as,

$$\mathbf{K}_V \mathbf{V} = \mathbf{F}_V, \text{ where } \mathbf{K}_V = \int_{\Omega} \sigma \mathbf{B}^T \mathbf{B} d\Omega, \text{ and } \mathbf{F}_V = \int_{\Gamma_j} \mathbf{N}^T j_n d\Gamma, \quad (2.19)$$

where $N(x)$ is the vector of shape functions $N(\mathbf{x})$, and $B = \nabla N(x)$. The quasi-static solution of above Eq. (2.21) is obtained during each time step using direct linear solver.

The discrete approximation of the weak form of the Eq. (2.19) is given as,

$$\mathbf{M}_T \dot{\mathbf{T}} + \mathbf{K}_T \mathbf{T} = \mathbf{F}_T, \quad (2.20)$$

where $\mathbf{T}(t)$ is the vector of nodal temperature, and

$$\mathbf{M}_T = \int_{\Omega} \rho_{eff} c_{eff} \mathbf{N}^T \mathbf{N} d\Omega, \quad (2.21)$$

$$\mathbf{K}_T = \int_{\Omega} k_{eff} \mathbf{B}^T \mathbf{B} d\Omega, \quad (2.22)$$

$$\mathbf{F}_T = \int_{\Omega} \mathbf{N}^T \dot{q}_{eff} d\Omega + \int_{\Gamma_q} \mathbf{N}^T q_n d\Gamma. \quad (2.23)$$

The solution of the nodal temperature at time $t + \Delta t$ is obtained using a 2-step Backwards Differentiation Formula (BDF) method [22].

Microscale solution approach—The EOS in Eq. (2.10) together with Clausius-Clapeyron equation (Eq. (2.7)) and the energy balance equation (Eq. (2.11)) describe the evolution of local thermodynamic states at each macroscopic spatial point $\mathbf{x} \in \Omega$. This gives a nonlinear set of equations, which is solved for the equilibrium temperature T using Newton's method. The modeling approach is summarized in Algorithm 2.1.

Algorithm 2.1**Solution algorithm**

```

1   Set  $t = 0$ ,  $x = 0$   $T_I^0 = T_{a'}$  and  $P = P_0$ 
2   for each time step  $t \rightarrow t + \Delta t$  until  $t < \tau$ 
3       Solve charge conservation law
4        $\mathbf{K}_V \mathbf{V} = \mathbf{F}_V$ 
5       Compute Joule heating
6        $\frac{\Delta q_e}{\Delta t} = \mathbf{J} \cdot \mathbf{E}$ , with  $\mathbf{J} = \sigma(T)\mathbf{E}$ 
7       for each macroscale Gauss point
8           if  $T > T_0$  then
9               Solve micromechanical model
10               $x = f(P, T)$ 
11               $h = q_e + v \cdot P$ 
12               $P = P_0 \exp \left[ -\frac{\lambda}{R_w} \left( \frac{1}{T} - \frac{1}{T_0} \right) \right]$ 
13          end if
14          Return effective properties:  $\rho_{eff}$ ,  $c_{eff}$ ,  $k_{eff}$ , and  $q_{eff} = \rho_{eff} c_{eff} T$ 
15      end for
16      Solve heat equation using 2-step BDF method
17  end for

```

3. Numerical examples

The two-scale model was first solved for a one dimensional problem of radio-frequency and then for a two dimensional axisymmetric model of electrosurgical ablation of liver tissue and the results were validated with *ex vivo* experimental data. The macroscale problem was solved using the commercial software package COMSOL Multiphysics® 5.2 (Comsol, Inc., Burlington, MA) with the microscale model being solved using MATLAB® R2015a (The MathWorks Inc., Natick, MA).

3.1. 1D radio-frequency heating

A transient one dimensional finite element computational model is developed to simulate radio-frequency heating of a 2 cm thick liver tissue with properties shown in Table 1. A normal current density of 0.05 A/mm² and thermal insulation is applied to the top surface of the liver. A ground condition is applied to the bottom surface which is assumed to be at a constant temperature of 37 °C. The tissue is initially at 37 °C and is heated for 5 seconds. The macroscopic domain along the 2 cm depth of the liver tissue is discretized using 400 linear finite elements with two Gauss points per element. The effective properties at each integration point are obtained by solving micromechanical model. Fig. 2 shows the

convergence of the tissue temperature in L^2 norm, when h -refinement is performed at the macroscale after 5 s of heating.

Next, the temperature distribution along the depth of the liver tissue is shown in Fig. 3 at different time intervals. Since the current density is constant along the depth, the heat generation rate is uniform resulting in a uniform temperature distribution along the depth except close to the 37 °C bottom boundary. Initially the temperature increase is rapid, however, once the temperature reaches 100°C, the temperature increase at a slower rate. This change in the rate of temperature increase for the two scale model can be clearly seen in Fig. 4(a) which shows the temperature variation with time at a depth $d = 0$ mm. The change in rate occurs at $t = 1.8$ s when the temperature reaches 100°C and the microscale model is activated to account for evaporation losses. The effect of the evaporation losses on temperature is significant when compared with the temperature variation with time of the macroscale heat diffusion model which does not account for evaporation, where the temperature exceeds 200°C after 5 s of heating. In the two scale model, once evaporation begins, the vapor fraction $\times$ begins to increase which can be seen in Fig. 4 (b) where the vapor fraction is plotted against time. At the end of the 5 s heating process 23.5% of the initial water content in the liver has evaporated.

3.2. Ablation of liver tissue

An axisymmetric finite element model of the liver tissue with dimensions 30 mm $\times$ 30 mm was developed and discretized using 4714 quadratic quadrilateral finite elements with four Gauss points per element. A ball-tip electrode was represented as a rigid body with tip radius $r_e = 1.17$ mm to accurately represent experimental conditions presented below. Prior to activation, the electrode was indented into the tissue by 1.8 mm to bring it into contact with the tissue as shown in Fig. 5. A graded mesh, as shown in Fig. 5, was used with a maximum element size of 2.085×10^{-4} m close to the electrode-tissue interface and a maximum element size of 1.327×10^{-3} m farther away.

Initial and boundary conditions—For the electrical problem, a normal current density was prescribed at the electrode-tissue contact region with the bottom of the tissue being grounded. For the energy conservation equation, the following initial and boundary conditions were prescribed:

$$T(\mathbf{x}, 0) = T_i \quad (3.1)$$

$$k_{eff} \nabla T(\Gamma_a, t) \cdot \mathbf{n} = h(T_{amb} - T(\Gamma_a, t)) + h_m x m (\rho_{v, amb} - \hat{\rho}_v(\Gamma_a, t)) (\lambda + c_w T_g(\Gamma_a, t)) \quad \text{on } \Gamma_a$$

$$(3.2)$$

$$\nabla T(\Gamma_i, t) \cdot \mathbf{n} = 0 \text{ on } \Gamma_i \quad (3.3)$$

where Γ_a is the boundary exposed to the atmosphere and Γ_i the insulated boundary with $\Gamma = \Gamma_a \cup \Gamma_i$ and $\Gamma_a \cap \Gamma_i = \emptyset$. $\mathbf{n}$ is the unit outward normal, $h = 3.1 \text{ W m}^{-2} \text{ K}^{-1}$ [25] is the heat transfer coefficient, and $h_m = 0.01 \text{ m s}^{-1}$ [26] is the mass transfer coefficient of vapor into air representing evaporation at the surface. The mean of the measured initial temperature T_i of the tissue was 10.6 °C, 12.9 °C, and 13.4 °C for 10, 25 and 50W, respectively.

Electrical, mechanical, and thermal properties—The properties of the tissue and electrode were obtained from literature and reported in Table 1. For the temperature range in which data are available for the temperature dependent electrical conductivity $\sigma(T)$ of the tissue, linear interpolation was used to compute its value at any intermediate point. Since no experimental data are available for the electrical conductivity beyond 100 °C, it is assumed to remain constant beyond this temperature. The reported value of the elastic modulus of liver ranges from 640 Pa to 1.1 MPa for strains <10% [27]-[33]. In this work, small deformation is assumed and the elastic modulus of liver was taken to be 1980 Pa [27]. The effect of using a large elastic modulus or a bilinear model as proposed by Majaron et al. [18] is briefly discussed in the results. The thermal contact resistance at the electrode–tissue interface was considered to be small $\sim 10^{-6} \text{ K m}^2 \text{ W}^{-1}$ such that,

$$k_{eff} \nabla T(\Gamma_e, t) \cdot \mathbf{n} = \frac{(T_e(\Gamma_e, t) - T(\Gamma_e, t))}{R_t} \text{ on } \Gamma_e \quad (3.4)$$

where Γ_e is the boundary where the tissue is in contact with the electrode, T_e is the electrode temperature at the boundary, and R_t is the thermal contact resistance.

Results—The simulation results were validated with electrosurgical ablation experiments performed on *ex vivo* porcine liver with a ball-tipped electrode. The details of the experiments are presented elsewhere [9] and will be briefly summarized here. Four samples of fresh porcine liver were locally sourced and kept in a static cold storage at 4 °C prior to start of the experiment. The samples were placed on a dispersive return pad (ESU Ground Pad 400–2100, Conmed Corporation, Utica, NY). An infrared (IR) camera (FLIR A655sc) with a 480p × 640p resolution and 30 Hz data acquisition rate was positioned vertically above the tissue and set to record surface temperature. A single image of a 10 mm × 10 mm square was taken to measure the pixel to mm scale for each measurement. An electrosurgical unit (System 2450® Electrosurgical Generator, Conmed Corporation) was used to perform a stationary ablation procedure on the liver using a 1.17 mm radius bent ball-tip electrode (Electrosurgical Ball Electrode A831, Bovie Medical Corporation, Clearwater, FL). Measurements were repeated twice on each liver at “pure cut” power setting 10W, 25W, and 50W, which is a pure sinusoidal signal at 400 kHz.

The experimental results from the infrared measurement of temperature are shown in Fig. 6. The infrared images show the temperature evolution for a single measurement at each of the

10W, 25W and 50W power settings, respectively. For the 10W and 25W power settings, the temperature distribution is found to be axis-symmetric for the measurements recorded. Some small variation however is observed in the 50W case which is attributed to the high power and speed of the ablation process where small variations in the tissue properties may cause lead to temperature asymmetry. To obtain the radial temperature variation with time from the infrared images, a region of interest around the electrode was considered. For each image, points that are radially equidistant from the center of the electrode were averaged to obtain the temperature at that radial distance and time. The length of the ablation process was limited by a sudden drop in the current and a corresponding increase in the voltage to an open circuit voltage that was observed shortly after the temperature reached 100°C near the electrode. The drop in current was attributed to possible desiccation and charring occurring at the electrode-tissue interface leading to an increase in the interfacial impedance.

The convergence in the L^2 norm of the relative error of temperature, when h -refinement is performed on the macroscopic domain is shown in Fig. 7. Temperature distribution as a function of radial distance from the electrode tip was recorded during the experiments as a mean of 8 measurements at four different time points corresponding to 10W, 25W and 50W power settings of the electrosurgical unit. The mean and standard deviation of each measurement is shown in Fig. 8 (a)–(c) with the variation of temperature (°C) plotted against the radial distance (mm) from the center of the electrode in solid lines. The corresponding simulation results are plotted as dashed lines. The results show that the maximum relative errors are 6.8 % ($t=4.662$ s), 9.7 % ($t=1.512$ s) and 8.7 % ($t=0.672$ s) corresponding to the 10W, 25W and 50W power settings, respectively. The evolution of temperature with time is plotted in Fig. 8 (d)–(f) for the 10W, 25W and 50W power settings at radial distances of 1.2 mm, 1.6, 2.0 and 3.0mm from the electrode-tissue interface. The maximum relative error for the 10 W, 25W and 50W cases are 6.6% ($R=1.6$ mm), 7.6% ($R=1.6$ mm), and 7.1% ($R=1.2$ mm), respectively. The maximum relative errors occur near the electrode and later in the ablation process, and may be attributed to the onset of vaporization below the boiling point of water and neglecting heat transfer due to transport and vapor diffusion within the tissue after evaporation. As the temperature increases and a gradient of the vapor fraction occurs due to evaporation within the tissue, some transport is expected and as time progresses transport effects can become more relevant. Since the model accounts for evaporation only after the temperature reaches the boiling point of water, the over prediction error in the 50W case could be attributed to surface evaporation losses occurring below 100 °C which are not adequately captured by the vapor diffusion boundary condition. Due to the higher power density in the 50W case, not accounting for evaporation losses below 100 °C would result in a more significant temperature increase compared to the 10W and 25W power settings. Once the temperature reaches 100 °C, the model is able to capture evaporation losses and prevent further increase in temperature.

The effect of losses due to phase change is further highlighted in Fig. 9 (a)–(c) which compare the temperature predictions of the two-scale model with that of a single-scale heat conduction model which does not account for phase change. Initially, the temperature prediction from both models coincide as expected, since the evaporation in the two-scale model does not occur below 100 °C. However, once the temperature exceeds 100 °C, neglecting evaporation results in over-prediction of the experimentally-observed temperature

at the end of the ablation process by approximately 7°C and 23°C near the electrode-tissue interface for the 25W and 50W cases, respectively. No over-prediction is observed in the 10 W case since less evaporation is expected at the lower power setting. Due to convection and heat loss through contact and vapor diffusion at the tissue surface, the effect of evaporation on the surface temperature is less pronounced than that observed in the 1D example in Fig. 4(a). However, the single scale model without evaporation predicts temperatures of 114°C, 159°C and 181°C for the 10W, 25W, and 50W cases, respectively at the hottest spot below the electrode, compared to 102–103°C predicted by the two-scale model. These differences in the temperature prediction can be attributed to the evaporation losses which differentiates the two-scale model from a standard heat diffusion model. The effect of evaporation losses is verified by considering the fraction of water content remaining in the tissue predicted in the model at $t=4.662s$, $t=1.512s$, and $t=0.672$ seconds for the 10 W, 25 W and 50 W power settings as shown in Fig. 9 (d)–(f) respectively. The largest evaporation loss occurs immediately below the electrode where the tissue is hottest. In this region, for the 10W power setting, 10% of the water is found to have been evaporated compared with 48% and 65% for the 25W and 50W power settings, respectively, where almost all of the water has evaporated, showing that the model can capture the evaporation losses. At the contact region there is little evaporation because of the thermal contact between the electrode and the tissue, where the electrode acts as a heat sink lowering the temperature to below 100 °C. Near the surface, the evaporation of water through the vapor diffusion boundary condition reduces the temperature and little water appears to have evaporated. This is because the model does not account for mass transport, so the water loss at the boundary does not affect the vapor fraction within the tissue.

To test the effect of tissue elasticity, a larger value of the elastic modulus of 750 kPa [26] was used in two cases. In one set of simulations, the higher elastic modulus was applied to the model throughout the heating process. In another set of simulations, a piece-wise model was used with the modulus of the liver increasing from 1.98 kPa to 750 kPa when the cavity expanded by 400% [18]. In both cases, with stiffer tissue and no transport, the pressure inside the tissue was observed to rise beyond 5 atm with very sharp pressure gradients near the electrode interface, with <1% water evaporation.

4. Conclusion

A two-scale model that comprises of the macroscale heat conduction coupled with microscale evaporation was described and applied to electrosurgical ablation procedures. Validation was done with mean experimental temperature distributions measured from electrosurgical ablation testing on *ex vivo* porcine liver at three different power settings. Simulation errors near the electrode-tissue interface for high power settings, particularly in the temperature range just below 100 °C, is attributed to the model not capturing vaporization losses below the boiling temperature of water. However, once the temperature exceeded 100 °C the microscale evaporation model was able to capture the resultant losses and prevent further significant increase in temperature, which agrees well with experimental observations. Significant errors observed below 100 °C further from the electrode at higher power settings indicate that the model can be further extended to include water-vapor mixture transport to accurately predict the thermal effects vapor diffusion through the tissue,

particularly at high power settings where most of the water below the electrode was found to have evaporated. For a complete description of the tissue degradation due to Joule heating and excessive pore pressure, a thermomechanical damage model needs to be employed. At the macroscale, a coupled electrical-thermal-mechanical analysis in conjunction with the thermomechanical tissue damage model will lead to a comprehensive model that can accurately predict tissue degradation in electrosurgical procedures.

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Fig. 1.
Schematic of two-scale model

Fig. 2.
Convergence of temperature when h-refinement is performed along the depth of the liver tissue

Fig. 3.
Temperature distribution at different times

Fig. 4.

(a) Variation of temperature with time at depth $d=0\text{mm}$ for the two scale model and the heat diffusion model with no evaporation and (b) variation of the vapor fraction x with time at depth $d = 0 \text{ mm}$.

Fig. 5.
Schematic of the computational model of liver tissue

Fig. 6.

Infrared images showing the evolution of temperature captured at different times for a single measurement at 10W (top row), 25W (middle row) and 50W (bottom row).

Fig. 7.

Convergence in L^2 norm of the relative error of temperature, when h -refinement is performed on the macroscopic domain

Fig. 8.

Comparison of mean experimental temperature to simulation results for the radial variation of temperature for (a) 10 W (N=8), (b) 25 W (N=8) and (c) 50W (N=8) and temperature evolution with time at four radial locations for (d) 10W, (e) 25W and (f) 50W.

Fig. 9.

(a) Comparison of temperature profiles predicted by the model and heat conduction with no evaporation for the (a) 10W, (b) 25W and (c) 50 W power settings, and fraction of water content remaining, with 1 indicating no evaporation occurred and 0 indicating all water has evaporated, as predicted by the model for (d) 10W power setting at $t=4.662s$, (e) 25W power setting at $t=1.512s$ and (f) 50 W power setting at $t=0.672s$

Table 1

Material properties used in the model

Tissue	k (W/m.K)	ρ (kg/m ³)	c (kJ/kg.K)	T (°C)	$\sigma(T)$ (S/m) ^{b,c}	E (Pa) ^d	ν ^{e,f}
Vapor ^a	0.016	Ideal Gas	2026	Liver 10	0.16	1980	0.48
Water ^a	0.57	998	4219	37	0.33		
Solid	0.28	1520	2607	80	0.79		
				90	0.75		
				100	0.64		
Electrode	k (W/m.K)	ρ (kg/m ³)	c (kJ/kg.K)				
Stainless steel ^a	11.9	7930	132				

^aCardarelli [23]
^bZurbuchen et al. [24]
^cGabriel et al. [28]
^dUmale et al. [27]
^eChui et al. [34]
^fKim et al. [35]