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A continuum thermomechanical model of *in vivo* electrosurgical heating of hydrated soft biological tissues

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Abstract

Radio-frequency (RF) heating of soft biological tissues during electrosurgical procedures is a fast process that involves phase change through evaporation and transport of intra- and extra-cellular water, and where variations in physical properties with temperature and water content play significant role. Accurately predicting and capturing these effects would improve the modeling of temperature change in the tissue allowing the development of improved instrument design and better understanding of tissue damage and necrosis. Previous models based on the Pennes' bioheat model neglect both evaporation and transport or consider evaporation through numerical correlations, however, do not account for changes in physical properties due to mass transport or phase change, nor capture the pressure increase due to evaporation within the tissue. While a porous media approach can capture the effects of evaporation, transport, pressure and changes in physical properties, the model assumes free diffusion of liquid and gas without a careful examination of assumptions on transport parameters in intact tissue resulting in significant under prediction of temperature. These different approaches have therefore been associated with errors in temperature prediction exceeding 20% when compared to experiments due to inaccuracies in capturing the effects of evaporation losses and transport. Here, we present a model of RF heating of hydrated soft tissue based on mixture theory where the multiphase nature of tissue is captured within a continuum thermomechanics framework, simultaneously considering the transport, deformation and phase change losses due to evaporation that occur during electrosurgical heating. The model predictions are validated against data obtained for *in vivo* ablation of porcine liver tissue at various power settings of the electrosurgical unit. The model is able to match the mean experimental temperature data with sharp gradients in the vicinity of the electrode during rapid low and high power ablation procedures with errors less than 7.9%. Additionally, the model is able to capture fast vaporization losses and the corresponding increase in pressure due to vapor buildup which have a significant effect on temperature prediction beyond 100 °C.

Keywords

electrosurgery; ablation; mixture theory; tissue heating

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1. Introduction

Accurate modeling of electrosurgical heating of soft tissues for interventional and therapeutic purposes requires a fundamental understanding of how the applied energy distributes and dissipates within the tissue. Electrosurgery consists of the application of alternating current in the radiofrequency range (300 kHz – 1 MHz) to simultaneously cut and coagulate tissues [1]. Tissue is rapidly heated to 100 °C within seconds [2], [3] when higher power settings are applied. This occurs in surgical procedures such as electrosurgical cutting, endometrial and prostate ablation at high power where a stationary probe is applied for a few seconds and then relocated to simultaneously vaporize and coagulate tissue [3], [4], liver resection [5] and radiofrequency-assisted liver resection where a pre-coagulation of tissue is performed for a few seconds before transection [6], [7]. In such electrosurgical procedures, the vicinity of the electrode is a region of high power density as the electric field intensity drops off as the square of the distance from it, resulting in sharp temperature and pressure gradients with large evaporation losses [8]. Computational models are often used to assess the safety of electrosurgical procedures, predict tissue damage, and to develop new instrument designs [9], but the model solutions often fail to capture experimental observations with accuracy. The challenge of accurately modeling RF electrosurgical heating of hydrated soft tissues arises from the need to capture energy dissipation that is significantly affected by changes in temperature dependent properties, latent heat loss, and phase change within the tissue that is heterogeneous in nature and contains more than 70% water by mass [10]–[13]. Modeling of energy dissipation during electrosurgery is often accomplished using the Pennes bioheat equation or a standard conduction equation [12]–[16]. While some works assume constant material properties, the temperature dependence of electrical conductivity which can grow by a factor of five from body temperature to 100 °C has been recognized by others such as Chen et al. [12], Chang [13] and Haemmerich and Schutt [14]. Yet, when predicted temperature is compared to experimental data, a maximum error of 23.9% was reported during *ex vivo* electrosurgical vessel sealing [12] and up to 27.4% during *ex vivo* radiofrequency ablation of normal porcine liver tissue [14]. The temperature dependence of the thermal properties and the latent heat loss associated with phase change are often neglected in most models of electrosurgical heating despite temperatures often exceeding 100 °C [1]. Previous work determined the apparent specific heat of *ex vivo* porcine liver by fitting to a transient heat transfer model that accounted for the temperature dependence of thermal and electrical properties. The error between simulation and experiment was minimized to a mean of 5.5% for 24 measurements, and a comparison between the variation of the apparent specific with temperature and the specific heat of porcine liver at body temperature was done. Results showed that the rate of energy losses due to latent heat and phase change was dominant at temperatures approaching 100 °C where it was larger than the rate of energy stored in the tissue by a factor of four [17]. This highlights the critical importance of accounting for phase change when modeling energy dissipation at temperatures above 100 °C.

Several approaches are found in the literature for modeling heating of biological tissues that include the effect of evaporation on energy dissipation in energy-based surgical procedures. Abraham et al. considered evaporation through a piece-wise description of the tissue

enthalpy with linear increase in the 99 – 100 °C range until all the water in the tissue has evaporated at 100 °C [18]. In other studies, such as the one by Yang et al., water content variation in ablated tissue was correlated to temperature and fitted to an approximate numerical model, noting that samples with residual water content exceeding 50% could not be measured due to a noticeable water drip in their method [19]. They used this correlation to include the effect of energy losses due to latent losses in a modified apparent specific heat, however simulation temperature results differed by more than 20 °C when compared with microwave heating of *ex vivo* bovine liver [20]. A mechanistic model proposed by Majaron et al. modeled the phase change during laser heating 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 which is equilibrated by increasing stresses in the surrounding solid until rupture [21]. This equation-of-state relating vapor mass fraction to temperature and pressure at the microscale was incorporated into a two-scale model to account for evaporation losses at the macroscale during electrosurgical ablation of *ex vivo* porcine liver tissue [22]. Comparison with experimental results showed that the model captured evaporation losses above 100 °C, however, the largest errors were observed farther from the electrode and later in the ablation process and were attributed to neglecting heat transfer due to vapor transport in the tissue after evaporation.

To account for the effect of transport in tissue, the multiphase nature of tissue has been considered in the literature using a porous media model for cooking processes. This model considers tissue to consist of a porous solid matrix with a largely interconnected void (pore) space occupied by a fluid. Volume averaging of the different phases at the microscale along with heat and mass conservation for each phase (solid, liquid and gas, consisting of air and vapor) is used to describe the heat and mass conservation of an average volume at the continuum macroscale [23], [24]. Scarpa and Milano used a simplified model to simulate phase change and adsorption in wet porous media by neglecting the mass fluxes of both the gas and liquid phase that rely on Darcy flow and permeability [25]. The porous media model including the mass transport of the gas and liquid phase was used to model the microwave heating of high and low moisture material by Ni et al [26]. The porous media model has been coupled with a linear elastic mechanics model to capture volume and pressure changes during rice puffing [27] and hyper elastic model by Dhall and Datta to capture shrinkage during drying and cooking of biomaterials [28]. One study performed *ex vivo* microwave ablation of liver tissue using an inserted antenna at 90W and 2.45 GHz during a 3 minute heating process and compared measured temperature with multiphase porous media model predictions [29]. While simulation offered a good match to testing on homogenized liver, under-prediction errors of up to 30 °C was reported when comparing with intact liver tissue, especially at higher temperatures and further from the antenna. These errors were attributed to inaccuracies in the assumptions on porosity and permeability and the corresponding properties which were estimated from foodstuff. The authors suggest that while water and vapor diffuse freely in homogenized liver, which is more similar to the assumed porous media environment, there is a larger resistance to diffusion in structurally and anatomically intact liver which was not accounted for in the model.

The significant discrepancies found between model predictions and experimental data in literature show that to accurately model radiofrequency heating of soft tissue, capturing the interactions between the heterogeneous phases that make up soft tissue including heat transfer, mass transport, phase change and mechanics as well as the careful examination of the underlying assumptions, is therefore critical. The heterogeneous nature of soft tissue and the interaction of phases at the microscale can be modeled as a homogeneous continuum at the macroscale through mixture theory [30], [31]. In the context of mixture theory, soft tissue can be treated as a mixture of phases, where each point in the mixture is simultaneously occupied by each constituent phase. For each constituent of the mixture, local forms of the mass, momentum and energy balance equations are satisfied, as well as for the mixture itself [31]–[34]. Satisfying the Second Law of Thermodynamics through the Clausius-Duhem inequality for the mixture can be used to obtain the general forms of the constitutive relations [35], [36]. This approach allows the incorporation of interactions between phases within an open system in a consistent framework.

Mixture theory has been widely applied to biological systems [37]–[40] including modeling of tumor growth [37], [38] where numerical simulations were able to capture experimentally observed complex tubular branching of an invasive tumor due to nutrient deprivation. It has also been applied to studying convection and diffusion in charged hydrated soft tissue where comparison between the convective and diffusive flux relative to the solid phase at low and high strain levels was found to be consistent with experiment [39]. The osmotic swelling of the annulus fibrosus of an intervertebral disc in a salt solution and subsequent confined compression and water expulsion was also modeled using mixture theory [40]. Simulation results were validated with experimental measurements of tissue height variation in time with a maximum relative error less than 2%. Mixture theory has also been used to model potato frying with inclusion of both water and oil phases in a viscoelastic solid, and both temperature and moisture content predictions were validated with experimental data [41]. The ability of mixture theory to predict complex behavior of soft tissue in different applications and match experiment indicates that it can offer a solid framework to model radiofrequency heating of hydrated soft tissue.

In this paper, we describe a model of electrosurgical tissue heating based on mixture theory with solid, liquid and gas phases, incorporating a non-equilibrium phase change model applicable to rapid vaporization processes. Mass, momentum and energy balance equations are described and constitutive relations are obtained from the Second Law of Thermodynamics. The predictions of this model are then compared with *in vivo* experimental data obtained from electrosurgical heating of porcine liver tissue.

The paper is organized as follows. Section 2 discusses the experimental procedure followed to collect power and temperature data during *in vivo* electrosurgical ablation tests of porcine liver tissue. Section 3 describes the mixture theory model for radio-frequency heating of soft tissues including the kinematics, mass, momentum and energy balance equations as well as constitutive relations obtained from the Clausius-Duhem inequality. The details of the numerical simulation of the experimental conditions are discussed in section 4. The experimental results and model predictions are presented and discussed in section 5 with concluding remarks in section 6.

2. Experimental Procedure

In vivo electrosurgical ablation experiments were performed to obtain power and temperature data for porcine liver tissue under an approved IACUC protocol at the Albany Medical College, Animal Resources Facility (Albany, NY). Ablation tests at different power settings (10, 25, and 50W) were repeated twice on different liver lobes of five pigs for a total of 10 measurements at 50W, and 9 measurements at 10W and 25W. The number of measurements was chosen based on a power analysis with G*power software [42] using a two-tailed t-test comparing experimental data to a one sample simulation prediction. While few studies were found that reported statistical power of the sample size for thermal measurement during ablation of tissue or tissue heating in general, in studies considering thermal or electrical measurements in soft tissue such as microwave ablation of *in vivo* porcine liver [43], radiofrequency ablation of *ex vivo* bovine liver [44], dielectric properties of normal and malignant *in vivo* human liver and *ex vivo* porcine liver tissue [45], and thermal heating effect of radiation on biological tissue [46], a sample size of 4 – 16 was considered sufficient to obtain statistical significance of mean values, with two studies [43], [44] showing a power of 90–95% for a sample size of 7–9. A statistical power level of 0.60 was considered to be sufficient in several studies on biological tissue properties [47]–[49] while a power level of 0.80 or greater is considered desirable [50]–[52]. Based on prior *ex vivo* testing and a pilot *in vivo* test with a significance level $\alpha=0.05$, a statistical power of $1-\beta=0.90$ is obtained with 9 measurements. The pigs, weighing 25–30 kg, were anaesthetized by injection of a cocktail of xylazine (2.2 mg/kg), ketamine (2.2 mg/kg), and telazol (4.4 mg/kg). Induction was done through inhalation of isoflurane at 5% and maintenance at 1.6 – 3%. Each pig was then placed in the supine position and a midline incision was used to open the abdomen. Depth of anesthesia was monitored by heart rate, respiratory rate, jaw tone and reflex response. Animal body temperature was monitored during the procedure and warming pads used as needed. Surgical personal protective equipment (PPE) was worn by veterinary staff and researchers throughout testing.

As shown in Figure 1, ablation experiments were performed on the exposed liver using a 1.17 mm radius bent ball-tip electrode (Electrosurgical Ball Electrode A831, Bovie Medical Corporation, Clearwater, FL) connected to an electrosurgical unit (System 2450 ® Electrosurgical Generator, Conmed Corporation). The electrode was held by a UR5 robotic arm (Universal Robots, Setauket, NY) at an angle to allow an unobstructed top view for the infrared (IR) camera. The ball-tip electrode was lowered until the tip was just at the tissue surface after exhale, then using the robot control, lowered into the tissue to obtain hemispherical contact. A dispersive return pad (ESU Ground Pad 400–2100, Conmed Corporation, Utica, NY) was placed securely on the thigh of the animal. An infrared camera (FLIR A655sc) with a 480pixel $\times$ 640pixel resolution and 30 Hz data acquisition rate was positioned above the liver to record surface temperature and aligned to be perpendicular with the ablation surface. Prior to testing, IR camera temperature measurement was verified to be within its specification accuracy through a supplier calibration certificate and independently checked by comparing with temperature measured using fiber optic temperature sensors (optical OTG-M280 temperature sensor with GaAs module, (Opsens Inc., Quebec City,

Canada) on liver in the range of 15 – 65 °C. The emissivity of the liver was assumed to be 0.9 and constant throughout the process [53].

A 400 kHz sinusoidal “pure cut” signal was applied at 10, 25 and 50W power settings. At each measurement, the electrode was placed in contact with the tissue and activated for 2 seconds. Electrode activation time was limited by the time between breaths to avoid errors in temperature recording due to breathing motion. Infrared video of the temperature field was acquired for a total of 30 seconds for each measurement. The voltage waveform generated by the electrosurgical unit (ESU) was measured using an oscilloscope (Agilent MSOX2024A) along with the RMS equivalent, and the current was found by measuring the voltage drop across a 10 Ω resistor placed in series with the electrode. A single image of a 16 mm $\times$ 16 mm square was taken to measure the pixel to mm scale for each measurement. The location of the electrode was identified from the infrared images. The radial temperature variation with time was obtained from the infrared images by considering a region of interest (ROI) around the electrode. In each infrared image frame, pixels in the ROI that are radially equidistant from the center of the electrode are averaged to obtain a mean temperature at a given radial distance and time.

3. Computational Model

In this section we present the model of electrosurgical heating of hydrated soft tissue based on mixture theory within a small deformation kinematics framework [30], [31]. Let Ω be an open bounded domain representing the tissue with boundary Γ . The tissue is assumed to be a tri-phasic mixture consisting of solid (s), liquid (l), and gas (g), that occupy the same material point $\mathbf{x} = \chi_{\alpha}(\mathbf{X}_{\alpha}, t)$ at time $t \in \mathbb{R}^+$ where $\mathbf{X}_{\alpha}$ denotes the reference point for the α^{th} phase ($\alpha = s, l \text{ or } g$) and χ_{α} the map from the reference to the current configuration as in Figure 2.

For a differential volume dv at $\mathbf{x}$, the portion occupied by phase α is denoted as dv_{α} . Thus the volume fraction of the α^{th} phase is defined as

$$\varphi_{\alpha} = \frac{dv_{\alpha}}{dv}, \quad (1)$$

with $\sum \varphi_{\alpha} = 1$.

If dm_{α} is the differential mass of phase α associated with dv_{α} , the “partial density” of phase α is defined as $\hat{\rho}_{\alpha} = dm_{\alpha}/dv$. Consequently, the mass density of the mixture is given as

$$\rho(\mathbf{x}, t) = \sum_{\alpha} \hat{\rho}_{\alpha}(\mathbf{x}, t). \quad (2)$$

The partial densities are related to the “intrinsic density” $\rho_{\alpha} = dm_{\alpha}/dv_{\alpha}$ as

$$\hat{\rho}_\alpha(\mathbf{x}, t) = \varphi_\alpha(\mathbf{x}, t) \rho_\alpha(\mathbf{x}, t). \quad (3)$$

Each phase is assumed to have the velocity

$$\mathbf{v}_\alpha(\mathbf{x}, t) = \frac{\partial \chi_\alpha(\mathbf{X}_\alpha, t)}{\partial t}. \quad (4)$$

The mixture velocity is given by the mass averaged velocity as

$$\mathbf{v}(\mathbf{x}, t) = \frac{1}{\rho(\mathbf{x}, t)} \sum_\alpha \hat{\rho}_\alpha(\mathbf{x}, t) \mathbf{v}_\alpha(\mathbf{x}, t). \quad (5)$$

3.1. Balance laws

Each phase in the mixture must satisfy its own balance laws which contain terms that represent the exchange of mass, momentum, and energy between the phases.

Mass Balance—We assume that mass exchange between the gas and liquid phases occurs due to phase change where $\pi_{\text{evap}}(\mathbf{x}, t)$ is the rate of evaporation ($\text{kg s}^{-1} \text{m}^{-3}$). Mass exchange for the solid phase is considered negligible and so is the change in partial density of the solid. Hence, the mass balance equations for the liquid and gas phases are given by

$$\frac{\partial \hat{\rho}_l(\mathbf{x}, t)}{\partial t} = -\pi_{\text{evap}}(\mathbf{x}, t) \quad (6)$$

and

$$\frac{\partial \hat{\rho}_g(\mathbf{x}, t)}{\partial t} + \nabla \cdot (\hat{\rho}_g(\mathbf{x}, t) \mathbf{v}_g(\mathbf{x}, t)) = \pi_{\text{evap}}(\mathbf{x}, t), \quad (7)$$

respectively.

It is reported in literature that blood perfusion slows down beyond 44 °C and is interrupted at temperatures > 50 °C due to coagulation [54]. Below 50 °C, the flow due to blood perfusion is assumed to be at steady state, thus the partial density of the liquid phase does not vary until evaporation begins to occur at 100 °C. Additionally, the estimated maximum possible rate of heat loss due to blood perfusion, given by $w_b c_b (T - T_b)$, is of the order of 10^5 W m^{-3} at 44 °C, considering a perfusion rate w_b of $26.6 \text{ kg m}^{-3} \text{s}^{-1}$ [55], specific heat of blood c_b of $3617 \text{ J kg}^{-1} \text{K}^{-1}$ and blood temperature T_b of 37 °C. In our experiments, high power fast ablation processes are considered in which the temperature may ramp up to

100 °C in less than 2s and, in literature, many electrosurgical procedures are performed in short bursts typically lasting less than 5 s [56], [57]. Thus, the rate of energy generation obtained from simulation in this work due to Joule heating is $\sim 10^7 - 10^{10} \text{ W m}^{-3}$ in the vicinity of the electrode, which is 2–5 orders of magnitude larger than the energy loss due to perfusion. In addition, the metabolic rate of liver tissue [58] reported in the literature is of order 10^5 W m^{-3} . Therefore, the effect of blood perfusion and the metabolic heat generation has been neglected in our analysis. For slow, low powered heating procedures in which 100 °C is reached in minutes rather than seconds, perfusion is expected to play a more significant role.

Momentum Balance—To get an order of magnitude estimation of the velocity of the liquid and gas phases due to pressure driven flow within the tissue, Darcy's law was applied assuming a pressure gradient of order 10^5 N m^{-3} as obtained from simulation results with a permeability of 10^{-17} m^2 [59] and 10^{-11} m^2 [60] for water and vapor respectively. Using Reynold's number for a packed medium [61], it was found to be of the order of 10^{-6} and 10^{-2} for water and vapor respectively. The compressibility of the gas phase given by $\beta = (1/\rho)(d\rho/dP)$, where ρ and P are the density and pressure of the gas estimated using steam tables, and found to be of order 10^{-6} [62]. The mass flux due to capillary diffusion is a function of the capillary pressure gradient in the material which can vary with concentration and temperature [24]. The diffusion due to capillary diffusivity within the tissue has been estimated using concentration gradients obtained from simulation and a mean diffusivity of $2.83 \times 10^{-10} \text{ m}^2 \text{ s}^{-1}$ [63] over a temperature range of 30–140°C. A maximum mass flux of $1.33 \times 10^{-3} \text{ kg m}^{-2} \text{ s}^{-1}$ and a Reynolds number of order 10^{-4} is obtained. Hence, the pressure driven and diffusive flow of the liquid phase was neglected and the corresponding momentum balance equation was not included in our analysis. The linear momentum balance of the gas phase reduces to Newtonian, incompressible, and creeping flow described by Darcy's law [64] as

$$\mathbf{v}_g(\mathbf{x}, t) = - \frac{\kappa_g}{\mu_g} \nabla P(\mathbf{x}, t) \quad (8)$$

where κ_g is the permeability of the tissue to the gas phase (m^2) and μ_g is the dynamic viscosity of the gas phase ($\text{Pa}\cdot\text{s}$), and $P(\text{Pa})$ is the total hydrostatic pressure. Justification for this equation is provided below (see Eq. 31).

Due to the small velocity of the fluid phases, the friction between the phases and viscous losses are negligible. We assume that there are no body forces acting on the solid (tissue). Hence, the linear momentum balance of the solid phase reduces to

$$\nabla \cdot \boldsymbol{\sigma}_s(\mathbf{x}, t) = \mathbf{0}, \quad (9)$$

Where $\boldsymbol{\sigma}_s$ is the Cauchy stress tensor. The constitutive relations are given in section 3.2.

Energy Balance—We assume that the solid, liquid and gas phases have the same temperature T locally at the same instant of time. To obtain an order of magnitude estimate of the time required to reach local equilibrium between the phases, a lumped capacitance approach was used to solve for two cube volumes in contact that are at a cellular scale, $20 \times 20 \times 20 \mu\text{m}^3$, at different initial temperatures to come to equilibrium. The time constant τ was found to be $\tau = \rho_1 c_1 L R / (1 + \rho_1 c_1 / \rho_2 c_2)$ where ρ and c are the density and specific heat respectively. Using properties for solid and liquid water provided below (see Table 1), and a thermal resistance R of $10^{-7} \text{ K m}^2 \text{ W}^{-1}$ estimated based on cell membrane thickness [10], [65], the time to equilibrium is $\sim 10^{-7} \text{ s}$.

The total mixture pressure P is the sum of the partial pressures of the three phases. In the absence of viscous heating as well as structural heating in the solid, the energy balance for the hydrated soft tissue in rate form for an open system as can be written as

$$(\rho c_p)_{eff} \frac{\partial T(\mathbf{x}, t)}{\partial t} = \dot{q}_e - \hat{\rho}_g(\mathbf{x}, t) \mathbf{v}_g(\mathbf{x}, t) \cdot \nabla (c_{p,g}(\mathbf{x}, t) T(\mathbf{x}, t)) + \frac{\partial T(\mathbf{x}, t)}{\partial t} - \nabla \cdot \mathbf{q} - \lambda(\mathbf{x}, t) \pi_{evap}(\mathbf{x}, t) \quad (10)$$

with

$$(\rho c_p)_{eff} = \hat{\rho}_s(\mathbf{x}, t) c_{p,s}(\mathbf{x}, t) + \hat{\rho}_l(\mathbf{x}, t) c_{p,l}(\mathbf{x}, t) + \hat{\rho}_g(\mathbf{x}, t) c_{p,g}(\mathbf{x}, t), \quad (11)$$

Where c_p is the specific heat capacity ($\text{J kg}^{-1} \text{ K}^{-1}$) and k is the thermal conductivity ($\text{W m}^{-1} \text{ K}^{-1}$) of the mixture, and λ is the specific enthalpy of evaporation (J kg^{-1}). The electrical generation term $\dot{q}_e (\text{W m}^{-3})$ is determined next, and $\mathbf{q}$ is the heat flux (W m^{-2}).

Electric Charge Conservation—The equivalent root mean square of the voltage and current, which are pure sinusoidal, are considered. Conservation of electrical charge provides the equation for the potential distribution $V(\text{V})$ in the tissue. The charge conservation in the tissue can be given as

$$\nabla \cdot \mathbf{J}(\mathbf{x}, t) = 0, \quad (12)$$

where $\mathbf{J} (\text{A/m}^2)$ is the current density. At the frequency used in electrosurgical procedures, capacitive losses are negligible and heat generation can be considered to be mainly resistive [12]. The current density can then be described by Ohm's law as

$$\mathbf{J}(\mathbf{x}, t) = s_e(\mathbf{x}, t) \mathbf{E}(\mathbf{x}, t), \quad (13)$$

where s_e is the temperature dependent electrical conductivity (S/m), and $\mathbf{E}$ is the electric field (V/m) given as

$$\mathbf{E}(\mathbf{x}, t) = \nabla V(\mathbf{x}, t). \quad (14)$$

The heat generated due to radiofrequency electrical excitation $\dot{q}_e (Wm^3)$ is

$$\dot{q}_e(\mathbf{x}, t) = \mathbf{J}(\mathbf{x}, t) \cdot \mathbf{E}(\mathbf{x}, t). \quad (15)$$

3.2. Constitutive Relations

We assume the following form of the specific free energy of the solid phase $\psi_s = \hat{\psi}_s(\boldsymbol{\varepsilon}, T)$, where $\boldsymbol{\varepsilon}$ denotes strain in the solid phase, and fluid phase $\psi_\alpha = \hat{\psi}_\alpha(\hat{\rho}_\alpha, T)$, for $\alpha = \{l, g\}$.

Invoking the classical Coleman and Noll arguments [35], for all admissible deformation and temperature fields, we can deduce the following constitutive relations that satisfy the Clausius-Duhem inequality representing the Second Law of Thermodynamics

$$\boldsymbol{\sigma}_s = \hat{\rho}_s \frac{\partial \hat{\psi}_s}{\partial \boldsymbol{\varepsilon}}, \quad (16)$$

$$\boldsymbol{\sigma}_g = -p_g \mathbf{I} + \boldsymbol{\tau}_g; \quad p_g = \hat{\rho}_g^2 \frac{\partial \hat{\psi}_g}{\partial \hat{\rho}_g} \equiv \varphi_g P, \quad \boldsymbol{\tau}_g = 2\mu \nabla \mathbf{v}_g, \quad (17)$$

$$\eta_\alpha = -\frac{\partial \hat{\psi}_\alpha}{\partial T} \text{ for all } \alpha = \{s, l, g\}, \quad (18)$$

$$\mathbf{q}_\alpha = -\mathbf{k}_\alpha \cdot \nabla T \text{ for all } \alpha = \{s, l, g\}, \quad (19)$$

where η_α is the specific entropy of phase α ($J \text{ kg}^{-1} \text{ K}^{-1}$), $\mathbf{q}_\alpha$ is the heat flux vector ($W \text{ m}^{-2}$), $\mathbf{k}_\alpha$ is a positive definite thermal conductivity tensor, and μ is the dynamic viscosity of the gas phase. Here, the partial stress of the gas (vapor) phase $\boldsymbol{\sigma}_g$ is split into partial hydrostatic pressure p_g and partial viscosity stress tensor $\boldsymbol{\tau}_g$.

Assuming diffusion of fluid through a solid undergoing small deformation with the solid strain defined as

$$\boldsymbol{\varepsilon} = \frac{1}{2}(\nabla u_s + \nabla u_s^T), \quad (20)$$

where u_s is the displacement of the solid. For the isotropic linear elastic solid, the free energy density functional can be given as

$$\hat{\rho}_s \psi_s = \frac{1}{2} \lambda_0 (\text{tr}(\boldsymbol{\varepsilon}))^2 + \mu_0 \text{tr}(\boldsymbol{\varepsilon}^2), \quad (21)$$

which applied to Eq. (16) gives us the constitutive relation for the Cauchy stress,

$$\boldsymbol{\sigma}_s = \lambda_s \text{tr}(\boldsymbol{\varepsilon}) \mathbf{I} + 2\mu_s \boldsymbol{\varepsilon} \quad (22)$$

where λ_s and μ_s are the Lamé coefficients. For the hydrated soft tissue, the solid stress can be decomposed into hydrostatic and deviatoric components as

$$\boldsymbol{\sigma}_s = -\varphi_s p \mathbf{I} + \boldsymbol{\tau}_s, \quad (23)$$

where the deviatoric stress $\boldsymbol{\tau}_s$ is defined as $\boldsymbol{\tau}_s = \boldsymbol{\sigma}_s - \frac{1}{3} \text{tr}(\boldsymbol{\sigma}_s) \mathbf{I}$.

For the gas phase, given the creeping incompressible Newtonian flow assumption and introducing an additive split of the stress tensor into the viscosity stress tensor and the hydrostatic pressure (Eq. (17)) and using the constitutive relation for $\boldsymbol{\tau}_g$, the linear momentum balance is then given by,

$$\mu_\alpha \nabla^2 \mathbf{v}_\alpha - \nabla p_\alpha = 0. \quad (24)$$

An assumption can be made that the viscous resisting force represented by the first term is linearly proportional to velocity in a porous medium with a creeping flow leading to Darcy's law in Eq. (8).

For the isotropic case, the thermal conductivity can be given by the partial thermal conductivity $\varphi_\alpha k_\alpha$ for phase α . The heat flux in the mixture is then given by,

$$\mathbf{q} = -k \cdot \nabla T, \quad (25)$$

where the mixture thermal conductivity k is,

$$k = \varphi_s k_s + \varphi_l k_l + \varphi_g k_g, \quad (26)$$

3.3 Evaporation

A non-equilibrium evaporation model which was first proposed by Bixler [66] is used to explicitly define the evaporation rate as

$$\pi_{evap} = K(p_{sat} - p_v), \quad (27)$$

where K is an empirical evaporation constant, p_{sat} is the saturation vapor pressure at temperature T , p_v is the vapor gas pressure obtained from solving the vapor mass balance and Darcy's law in Eqs. (8) and (9). The evaporation model is activated when the temperature exceeds 100 °C. The value of K was chosen to be sufficiently large for the evaporation rate to be near equilibrium with a value of 1.5×10^4 m·s. Too large values of K prevented the solution from converging and a constraint was placed on the total evaporation such that the maximum evaporation losses cannot exceed the total amount of Joules heating applied to the tissue. Thus, for the lower power setting of 10W a smaller evaporation constant 5×10^3 m·s was used to remain within the stated constraint and ensure convergence of the solution.

3.4. Boundary and Initial conditions

The part of the boundary (Γ) exposed to the atmosphere is denoted as Γ_a . The boundary far from the electrode with no temperature variation was considered insulated and denoted as $\Gamma = \Gamma_a \cup \Gamma_i$ with $\Gamma = \Gamma_a \cup \Gamma_i$ and $\Gamma = \Gamma_a \cap \Gamma_i = \emptyset$. $\mathbf{n}$ is the unit outward normal to the boundary and $\mathbf{u}_s$ is the displacement of the solid. Γ_e denotes the boundary in contact with the electrode and Γ_g is the boundary in contact with the ground. The initial conditions (I.C.) and boundary conditions (B.C.) for mass, momentum and energy balance equations for the solid, liquid, and gas phases are therefore given as:

Mass Balance

$$\text{Liquid} \quad \text{I.C.} \quad \hat{\rho}_l(\mathbf{x}, 0) = \varphi_{l,0} \rho_l \quad (28)$$

$$\text{Gas} \quad \text{I.C.} \quad \hat{\rho}_g(\mathbf{x}, 0) = \varphi_{g,0} \rho_g \quad (29)$$

$$\text{B.C.} \quad \hat{\rho}_g \mathbf{v}_g(\Gamma_a, t) \cdot \mathbf{n} = -\hat{\rho}_g(\Gamma_a, t) \frac{\kappa_g}{\mu_g} \nabla P(\Gamma_a, t) \cdot \mathbf{n} + h_m \varphi_g (\rho_g(\Gamma_a, t) - p_{v,amb}) \quad \text{on } \Gamma_a \quad (30)$$

$$\hat{\rho}_g(\Gamma_i, t) = \varphi_{g,0} \rho_g \quad \text{on } \Gamma_i \quad (31)$$

For the mass balance, the tissue is assume to have an initial volume fraction of liquid $\varphi_{l,0}$ and of gas $\varphi_{g,0}$. At the boundary Γ_a the gas mass flux out of the tissue is due to pressure driven flow and mass diffusion to the atmosphere.

Momentum Balance

$$\text{Solid} \quad \text{B.C.} \quad \boldsymbol{\sigma}_s(\Gamma_a, t) \cdot \mathbf{n} = \mathbf{0} \quad \text{on } \Gamma_a \quad (32)$$

$$\mathbf{u}_s(\Gamma_i, t) = \mathbf{0} \quad \text{on } \Gamma_i \quad (33)$$

$$\text{Gas} \quad \text{I.C.} \quad P(\mathbf{x}, 0) = p_{atm} \quad (34)$$

$$\text{B.C.} \quad P(\Gamma, t) = p_{atm} \quad \text{on } \Gamma \quad (35)$$

Traction-free boundary conditions were applied to the surface of the tissue, while zero displacement boundary condition was applied on Γ_i . The initial pressure in the tissue and at the boundary Γ is assumed to be at atmospheric pressure.

Energy

$$\text{Mixture} \quad \text{I.C.} \quad T(\mathbf{x}, 0) = T_0 \quad (36)$$

$$\text{B.C.} \quad -k \nabla T(\Gamma_a, t) \cdot \mathbf{n} = h_c(T(\Gamma_a, t) - T_{amb}) + c_{p,g}(\Gamma_a, t)T(\Gamma_a, t) \quad (37)$$

$$\left(-\hat{\rho}_g(\Gamma_a, t) \frac{\kappa_g}{\mu_g} \nabla P(\Gamma_a, t) \cdot \mathbf{n} = h_m(\rho_g(\Gamma_a, t) - p_{v,amb}) \right) \quad \text{on } \Gamma_a$$

$$\nabla T(\Gamma_i, t) \cdot \mathbf{n} = 0 \quad \text{on } \Gamma_i \quad (38)$$

The tissue is assumed to be initially at a temperature $T_0 = 37^\circ\text{C}$. Along the boundary exposed to the atmosphere, heat flux leaving the tissue is due to convection and latent losses due to pressure driven flow and diffusion of vapor to the surrounding air.

Electrical

$$\text{B.C.} \quad \mathbf{n} \cdot \mathbf{J} = j_n \text{ on } \Gamma_e \quad (39)$$

$$V = 0 \text{ on } \Gamma_g \quad (40)$$

A constant current density is applied at the tissue boundary in contact with the electrode and a zero potential at ground.

4. Numerical simulation

The computational model was implemented and solved using COMSOL Multiphysics® 5.2 (Comsol Inc., Burlington, MA). A 2D axisymmetric finite element model was developed to solve the electric field, mass, momentum and energy balance equations, respectively. Prior to heating, the electrode, which is initially at the surface of the tissue is displaced downwards by 1.8 mm to bring it into contact with the tissue. A linear elastic model is used to solve for the tissue deformation. Figure 3 shows the model mesh and the contact region of the tissue with the electrode.

The boundary conditions applied on the tissue are as described previously in Eqs. (35)–(47). The electrode is assumed to be initially at 20°C . The top of the electrode is assumed to be at a constant temperature 20°C consistent with IR camera measurements obtained during experiment, and the side to have a convective boundary condition with a calculated natural convection coefficient of $20.25 \text{ W m}^{-2} \text{ K}^{-1}$ as shown in Figure 3. The thermal contact resistance between the tissue and electrode was estimated based on a Mikic correlation [67], [68] to be $1 \times 10^{-6} \text{ K m}^2 \text{ W}^{-1}$. The properties of the solid, water and vapor of the tissue as well as other parameters used in the model are shown in Table 1. The elastic modulus of liver was found to vary widely in the literature with values ranging from 640 Pa to 1.1 MPa for strains $<10\%$ [69]–[72] and is dependent on the measurement method. In this work, we have used a value of 1980 Pa [69], since small deformation is assumed. During heating, biomolecular and structural changes may occur, e.g., protein denaturation. The effects of these changes have been incorporated in the model through the temperature dependence of tissue properties. The electrical conductivity $s_e(T)$ of liver tissue was obtained from literature, see Table 2. During simulations, linear interpolation was used to compute $s_e(T)$ up to 100°C , beyond which it was assumed constant.

5. Results and Discussion

The results of the electrical and measurements as well model validation with the *in vivo* ablation experiments are presented next. Experimental results and a validation of the model with ablation and volumetric heating experiments on *ex vivo* porcine liver are available in supplementary document to this work.

5.1. Electrical measurements

The measured root mean square current and voltage are shown in Figure 4 (a) and (b), respectively for a single ablation test at 10W, 25W and 50W. While the current remained constant during the 10W ablation for the total activation time of 2 s, at the higher power settings, even though the electrode was activated for the total time of 2 s, the measured current applied to the tissue was interrupted, dropping abruptly to near zero while the corresponding voltage across the electrode and tissue increased to open circuit voltage. This was attributed the rapid heating where there was increased resistance at the interface due to desiccation and possible charring causing the current to drop. Thus the ablation time was self-limited at the higher power settings. The mean results of the electrical measurements are shown in Table 3, where the mean root mean square current and the mean total voltage across the electrode and tissue were obtained from the mean of the measured values during each ablation test prior to the current drop with the standard deviation shown in parenthesis. The total power applied to the tissue remained constant and within 3% of the nominal power setting.

5.2. Temperature measurement and modeling

Examples of temperature measurements at different times during the *in vivo* porcine studies are shown in Figure 5 for the 10 W, 25 W, and 50 W power settings of the ESU. The cold zone which appears in the infrared images adjacent to the electrode is the bent ball-tip electrode above the tissue which partially obstructs the infrared camera view as shown in Figure 6. This region is excluded when obtaining the temperature variation in the tissue. The temperature distribution is found to be axisymmetric for the 10W case and at the initial time steps for the 25 W and 50 W cases. Later in the ablation process, small local hot spots that appear and then dissipate can be seen in the 25 W and 50 W cases, as well as some temperature asymmetry particularly in the 50 W case which may be due to the high power and speed of the ablation process combined with small variations in tissue properties and local water content.

The mean experimental radial temperature distributions for the *in vivo* porcine experiments are shown in Figure 7(a)–(c) for the 10, 25 and 50W cases respectively with the interquartile range in error bars. The computational results capture the temperature field with a maximum relative error norm of 2.7%, 3.8%, and 4.8% for the 10 W, 25 W, and 50 W cases, respectively. The variation of temperature with time is plotted in Figure 7 (d)–(f) for the 10W, 25W and 50W power settings, respectively, at radial distances of 1.2, 1.6, 2.0, and 3.0mm from the center of the electrode. The maximum relative error norm for the 10 W, 25W and 50W cases are 3.7%, 5.5%, and 5.8%, respectively at a radial distance of 1.2mm from the electrode for the 10W, 1.6mm for the 50W case, and at 2.0mm for the 25W case.

The larger errors are closer to the electrode and at the temperature peak during the ablation process for all power settings. The error is attributed to electrode-tissue contact properties that may change with tissue coagulation and desiccation, as well as evaporation occurring below 100 °C which is not captured by the evaporation model. Once the evaporation model is activated above 100 °C, further increase in temperature is prevented as seen in Figure 7 (d)–(f) at $R=1.2\text{mm}$ radial distance for all three power settings. The cooling phase of the temperature variation during the 2 seconds of temperature recording is also shown in Fig. 6. In model prediction of the temperature, the current applied to the tissue in simulation is reduced to 0.022 A at 0.8s and 0.026 A at 0.4s as measured from experiment for the 25W and 50W cases, respectively. In addition, some loss of contact is expected to occur at the electrode-tissue interface once the current drops. This loss of contact, most likely due to due to vaporization, desiccation and charring of tissue, results in an increased thermal contact resistance [80]. This is also supported by the large increase observed in electrical contact resistance as obtained from voltage and current measurements. Specifically, total resistance seen by the electrosurgical unit increases from approximately 300 Ω to over 13,000 Ω once the current drops. While not measured directly, electrode tissue thermal contact was previously reported to vary from 1.5×10^{-6} to $1 \text{ K m}^2 \text{ W}^{-1}$ [81]. Thus, after the current drop, the thermal contact resistance was set to $1 \text{ K m}^2 \text{ W}^{-1}$ for both power settings. Results show that while the model is able to capture the temperature decrease in the tissue within the time frame shown, the predicted temperature decreases slightly faster than in experiment. This may be due to irreversible changes in tissue properties and in electrode-tissue contact that would occur with continued thermal damage particularly at higher power settings.

The temperature distribution predicted by the model in the vicinity of the electrode and tissue interface is shown in Figure 8(a)–(c) at the end of the ablation process for each power setting. The hottest spot was found to occur in the tissue just underneath the electrode reaching about 102 °C for the 10 W and 25W cases and 105 °C for the 50 W case. This corresponds to the significant evaporation losses observed in Figure 8(d)–(f). The largest evaporation losses occur in the hottest spot underneath the electrode where for the 10 W case approximately 75% of the water has evaporated, while at the surface the maximum water loss is 21% near the electrode. In the 25 W case, 96% of the water is found to have evaporated in the region below the electrode with a maximum of 21% of water loss at the surface. Despite the increased power, the relatively modest increase in evaporation observed in the 25 W case compared to the 10 W case is due to the significantly smaller heating time after reaching 100 °C. Further increase in power density in the 50W case however, is large enough that despite the short heating time of 0.384 s, the water is rapidly vaporized underneath the electrode as shown in Figure 8(f). At the surface, the maximum water loss is 33%. The water loss at the surface for all three power settings falls in the range of 19–47% which is consistent with what is reported in literature for experimentally measured water loss after heating muscle tissue up to an internal temperature of 90°C [82]–[84]. The higher water loss predicted under the electrode for the 25W and 50W cases is consistent with the abrupt drop in the measured current (Figure 4) attributable to dry charring. These results show the ability of the non-equilibrium evaporation model used to capture large and rapid evaporation losses during high power radiofrequency heating of hydrated soft tissue.

The vaporization of the water in the tissue causes an increase in the fluid pressure which can be seen in Figure 9 (a)–(c) for the different power settings. Corresponding to the same maximum temperature, less pressure accumulation and a smaller vapor content is observed in the 10 W case compared to the 25 W case. This is due to the slower heating process in the 10 W case in which the vapor has more time to diffuse through the tissue resulting in lower pressure. The larger evaporation losses that occur in the 50W case coupled with the faster heating process means that vapor is unable to escape quickly enough via diffusion to the surface. Thus ablation at the higher power setting results in a significant pressure buildup during rapid heating. If heating is too rapid for the excess pressure to dissipate, e.g., when there is very high power density, pockets with vapor accumulation may occur followed by the rupturing of tissue and dissipation of the heat and pressure.

The effect of the evaporation constant on the solution of the model was considered for the 50W case. In Figure 10 temperature variation with time for different values of the evaporation constant is shown. Results show that when evaporation losses are completely removed, there is a significant over prediction of the temperature by up to 75°C near the electrode. Setting the evaporation constant to zero, also removes the effect of vapor transport in the tissue where there is no increase in pressure due to evaporation, and no pressure driven flow. Once evaporation losses are included, this maximum over prediction near the electrode is reduced for an evaporation constant of 500 to 35°C. Further reduction in temperature prediction over 100°C occurs when the evaporation constants are increased to 5×10^3 and 1.5×10^4 , allowing the model to capture larger evaporative losses. The small difference in temperature prediction between the two larger evaporation constants suggest that at a value of $K = 1.5 \times 10^4$, the evaporation rate is close to the equilibrium evaporation rate. Far from the electrode where no evaporation losses occur, the temperature distribution is not affected by the evaporation constant.

6. Conclusion

In this paper, a model based on continuum mixture theory that accounts for different phases within an open system was described and applied to radiofrequency electrosurgical procedures. Validation was done with mean experimental temperature distributions measured from ablation of *in vivo* porcine liver. Comparisons between experiment and model showed a maximum prediction error of 7.9%, with considerably more accurate predictions for low power setting and at the initial time steps of the higher power settings. At higher temperatures in the vicinity of the electrode, the model over-predicted the temperature as it approached 100 °C. However, once the temperature exceeded the boiling point of water, the evaporation model was able to capture vaporization losses and prevent further increase in temperature. The evaporation losses were consistent with pressure increase within the tissue and matched observations of desiccation and charring at higher powers in the experiment. These were also consistent with the electrical measurements showing a drop in current due to increased resistance at the interface.

The physical parameters used including the vapor diffusion coefficient at the boundary or the permeability of liver tissue, are mean values or estimates obtained for liver from available literature. Improving the accuracy of these physical parameters, which could have

dependence on water content or temperature, could improve the prediction of surface evaporation losses below 100 °C and further reduce the error found near the electrode-tissue interface at higher power settings. Another limitation of this study is that the model assumes a linear elastic tissue behavior. The incorporation of a finite deformation hyper-elastic model would more accurately represent the physical deformation of soft tissue and further improve the proposed mixture model. Future addition of a damage model along with a more detailed description of microscale effects such as protein denaturation coupled with the mixture model would offer a more complete description of tissue degradation and damage due to thermomechanical effects. However, further experiments are necessary to develop such a damage model. However, overall, the current model presents a rigorous approach for taking into account multiple thermal transport and phase change phenomena along with deformation for accurate prediction of temperature rise during RF electrosurgical heating.

Supplementary Material

Refer to Web version on PubMed Central for supplementary material.

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Research Highlights:

- A continuum thermomechanical model of radio-frequency electrosurgical tissue heating that is based on mixture theory is proposed.
- Model accounts for multiphase nature of tissue including a non-equilibrium phase change model, and satisfies mass, momentum and energy balance with constitutive relations obtained from second law.
- Modeling results are validated with experimental data obtained during electrosurgical ablation of *in vivo* porcine liver.

Figure 1.
Schematic of the *in vivo* experimental setup

Figure 2.
Kinematics of radiofrequency activated soft tissue.

Figure 3.
Finite element model of the computational domain.

Figure 4. Measured root mean square (a) current and (b) voltage for a sample measurement at 10W, 25W and 50W

Figure 5.

Infrared images showing the evolution of temperature in the *in vivo* porcine experiments, captured at different times for sample measurements at 10W (top row), 25W (middle row) and 50W (bottom row).

Figure 6.
Photographs showing electrode placement and resulting camera view occlusion.

Figure 7.

Variation in mean experimental temperature (solid lines) for in the *in vivo* experiments compared to simulation results (dashed lines) with radial distance for (a) 10 W, (b) 25 W, (c) 50W, and with time for (d) 10 W, (e) 25 W and (f) 50W.

Figure 8.

Temperature distribution in the electrode and tissue at (a) 2.160 s for 10 W, (b) 0.721 s for 25 W, and (c) 0.384 s for 50 W cases. Ratio of vapor content to initial water content (d) at 2.034 s for 10 W (e) at 0.721 s for 25 W and (f) at 0.384 s for 50 W cases.

Figure 9.

Pressure distribution in the tissue at (a) 2.034 s for 10 W (b) 0.721 s for 25 W and (c) at 0.384 s for 50 W cases.

Figure 10.
Variation of temporal temperature distribution in the 50W model with different values of evaporation constant K .

Table 1

List of material parameters used in model

Parameter		Value	Unit	Source
Density				
Liver	ρ_{liver}	1060	kg m^{-3}	[10]
Water	ρ_w	998	kg m^{-3}	[73]
Vapor	ρ_v	$\frac{p}{R_w T}, R_w = 461.5 \text{ J kg}^{-1} \text{ K}^{-1}$	kg m^{-3}	
Solid	ρ_s	1520	kg m^{-3}	Back calculated from liver [10]
Specific heat capacity				
Liver	$c_{p,liver}$	3700 at 37°C	$\text{J kg}^{-1} \text{ K}^{-1}$	[10]
Water	$c_{p,w}$	4219	$\text{J kg}^{-1} \text{ K}^{-1}$	[73]
Vapor	$c_{p,v}$	2026	$\text{J kg}^{-1} \text{ K}^{-1}$	[73]
Solid	$c_{p,s}$	3523	$\text{J kg}^{-1} \text{ K}^{-1}$	Back calculated from liver [10]
Thermal conductivity				
Liver	k_{liver}	0.518 at 37°C	$\text{J kg}^{-1} \text{ K}^{-1}$	[10]
Water	k_w	0.57	$\text{W m}^{-1} \text{ K}^{-1}$	[73]
Vapor	k_v	0.016	$\text{W m}^{-1} \text{ K}^{-1}$	[73]
Solid	k_s	0.28	$\text{W m}^{-1} \text{ K}^{-1}$	Back calculated from liver [10]
Permeability of gas phase in tissue	k_g	4.57×10^{-11}	m^2	[10]
Dynamic viscosity of vapor	μ_g	1.8×10^{-5}	Pa.s	[73]
Enthalpy of evaporation	λ	2.26×10^6	J kg^{-1}	[10]
Initial volume fraction of liquid phase	$\Phi_{l,0}$	0.74		0.7 mass fraction of water in liver [10]
Initial volume fraction of gas phase	$\Phi_{g,0}$	$0.1\Phi_{l,0}$		
Convection coefficient	h_c	3.1	$\text{W m}^{-2} \text{ K}^{-1}$	[74]
Mass transfer coefficient	h_m	0.01	m s^{-1}	[75]
Evaporation constant	K	$1.5 \times 10^4 (5 \times 10^3 \text{ at } 10\text{W})$	m s	
Elastic modulus	E	1980	Pa	[69]
Poisson's ratio	ν	0.48		[65][66]

Table 2

Temperature dependent electrical conductivity of liver tissue [67][68]

Temperature, T(°C)	Electrical conductivity, s_e (S)
10	0.16
37	0.33
80	0.79
90	0.75
100	0.64

Table 3

Mean electrical measurements for *in vivo* ablation experiments with standard deviation within parenthesis

Power setting	Mean RMS current (A)	Mean RMS voltage (V)	Power applied (W)
10 W	0.184 (0.008)	56.2 (2.8)	10.20 (0.31)
25 W	0.294 (0.013)	87.1 (4.7)	25.23 (0.47)
50 W	0.427 (0.023)	121.3 (9.8)	51.00 (2.49)