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Waveform-Dependent Electrosurgical Effects on Soft Hydrated Tissues

Electrosurgical procedures are ubiquitously used in surgery. The commonly used power modes, including the coagulation and blend modes, utilize nonsinusoidal or modulated current waveforms. For the same power setting, the coagulation, blend, and pure cutting modes have different heating and thermal damage outcomes due to the frequency dependence of electrical conductivity of soft hydrated tissues. In this paper, we propose a multi-physics model of soft tissues to account for the effects of multifrequency electrosurgical power modes within the framework of a continuum thermomechanical model based on mixture theory. Electrical and frequency spectrum results from different power modes at low- and high-power settings are presented. Model predictions are compared with in vivo electrosurgical heating experiments on porcine liver tissue. The accuracy of the model in predicting experimentally observed temperature profiles is found to be overall greater when frequency-dependence is included. An Arrhenius type model indicates that more tissue damage is correlated with larger duty cycles in multifrequency modes.

[DOI: 10.1115/1.4042898]

Keywords: electrosurgery, waveform, multifrequency, soft tissue heating

1 Introduction

Electrosurgery is one of the most common surgical techniques accounting for more than 80% of all cutting and coagulations procedures in laparoscopic surgery [1,2]. Electrosurgical units (ESU) provide an alternating current in the frequency range of 300 kHz–1 MHz to generate heat through conductivity losses in the tissue. Different power modes, shown in Fig. 1, are used for the purpose of producing specific cutting and coagulation effects [3]. The “pure” or “cut” waveform consists of a pure sinusoidal signal at a single nominal frequency in the radiofrequency range, while a “blend” waveform is a modulated pure sinusoidal waveform at 50–80% duty cycle with a similar fundamental frequency and a smaller modulation frequency around 25 kHz. The “coagulation” waveform is a short high-voltage radiofrequency waveform followed by a long rest resulting in a low duty cycle in the range of 5–40% and a modulation frequency ranging from 20 to 50 kHz [4,5]. The different power modes or waveforms produced by ESUs are utilized in different surgical applications. Pure and blend modes are typically used in resection of different organs such as the liver and uterus [6,7]. Contact desiccation is performed using coagulation modes with a stationary rollerball electrode [6], with spray coagulation used to coagulate and achieve hemostasis [8] and in electrosurgical vaporization procedures such as treating bladder outflow obstructions [9].

Only a few studies may be found in the literature that have examined the clinical and physical effects of using different waveforms during surgical procedures and their relation to the power applied. Chino et al. found that increased tissue injury depth in polypectomy correlated with coagulation waveforms compared to blend or pure cut at the same applied power [10]. Blend mode was

recommended since it enhanced hemostasis without a large increase in thermal damage. Tucker et al. showed that using the coagulation waveform on a rat uterine model for desiccation resulted in loss of tissue at the site 12 weeks postsurgery even at low-power settings [11]. A study by Fry et al. on patients undergoing polypectomy during colonoscopy compared the damage due to electrosurgery and histologic diagnostic quality when using a blend mode at 30 W setting or automatically switching between cutting and coagulation modes at 60 W. The results showed that polyps removed with the blend power mode had less diagnostic quality and slightly larger electrosurgery-related damage despite the lower power setting used [12]. These results show that at similar power settings, a difference in the tissue thermal damage and surgical end results are observed when using different waveform settings, yet conclusions over which power mode causes more damage varies between studies. With the lack of definitive guidelines, surgeons use a variety of settings. A survey of 478 surgeons showed that 74% utilize the coagulation mode most frequently during laparoscopic surgery. One-third of the total number of surgeons surveyed routinely used high-voltage and high-power (>40 W) coagulation and blend power modes while the remaining two-thirds routinely employed 20–40 W of power. The use of coagulation modes, which despite their low duty cycle, have significantly higher peak-to-peak voltage than cut or blended modes, increases the risk of capacitive coupling and directly increases the distance an electrosurgical spark travels to nearby tissues or instruments, thus increasing the risk of electrosurgical burns or complications. Of the surgeons surveyed, 18% reported at least one experience with electrosurgical burns during patient procedures [2]. An accurate physics-based model of electrosurgery can help develop guidelines on the power modes and settings needed for specific tissue outcome. Such guidelines could reduce the possible risk of complication by avoiding high-voltage modes when they are not needed.

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Manuscript received June 11, 2018; final manuscript received February 12, 2019; published online March 25, 2019. Assoc. Editor: Ram Devireddy.

Fig. 1 Summary of the waveform, duty cycle, and modulation frequency of different electro-surgical power modes (*top*: pure cut, *middle*: blend, and *bottom*: coagulation)

Accurate characterization of tissue effects based on power modes is problematic as the blend and coagulation modes employ modulated or nonsinusoidal waveforms with relatively broad frequency spectra. As heating in soft tissues occurs due to conductivity losses, and the electrical conductivity of tissue is dependent on both frequency [13–16] and temperature [17,18], accurately predicting losses in the tissue requires accounting for these multiple frequency components in the waveforms. Though significant work exists for the modeling of tissue heating due to pure sinusoidal waveforms [19–26], to the best of our knowledge, there are no studies that model the effect of modulated multifrequency waveforms applied to the tissue during electrosurgery. However, modeling the effects of each waveform of a multifrequency signal on temperature would clarify the contribution of power and frequency to the differences in heating observed, thus allowing for more accurate prediction of tissue damage and tailoring devices and waveforms to specific surgical tasks. In this paper, we determine the contribution of power and frequency on multifrequency tissue heating and develop a frequency-dependent waveform model to accurately capture the power dissipation from nonsinusoidal modulated signals by accounting for electrical property variation with frequency.

Another complexity in modeling radiofrequency electrosurgical heating in soft hydrated tissues is the need to account for energy dissipation that is significantly affected by changes in temperature-dependent tissue properties, latent heat loss, and phase change [21,27]. 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 [23]. In addition, the sensitivity of the temperature variation to the apparent specific heat, which included the effect of evaporation losses, as well as the sensitivity to the electrical conductivity and thermal conductivity, showed that temperature distribution was highly sensitive to the apparent heat storage and electrical conductivity losses in the tissue. Hence, in this paper the electrical model is coupled to a thermomechanical model of tissue heating based on mixture theory to predict the temperature variation in the tissue during electrosurgery [20]. The electrical waveform measurements and frequency spectrum results are discussed and compared for three power modes at low- and high-power settings. The frequency-dependent model results are then compared with those obtained assuming a single frequency model of electrical conductivity. Model predictions are compared with temperature measurements during in vivo electrosurgical heating tests on porcine liver at different power settings.

The paper is organized as follows. The multiphysics mathematical model of electrosurgical heating of soft tissue using a multifrequency power source is presented in Sec. 2. The results are validated with mean experimental temperature distributions measured from electrosurgical testing on in vivo porcine liver at different power settings in Sec. 3, followed by conclusions and future scope of work in Sec. 4.

2 Mathematical Model

In this section, we present a multiphysics tissue model for electrosurgery using a multifrequency power source. The

Fig. 2 Schematic of tissue domain and boundary

frequency-dependent electrical problem is presented in Sec. 2.1 to obtain the power dissipation at different frequencies in the tissue. The mass, momentum, and energy balance equations corresponding to a continuum mixture theory model of the tissue are presented in Sec. 2.2. Discretized equations are presented in Sec. 2.3.

2.1 Frequency-Dependent Model. In this section, the electrical problem is first defined in the time domain. A Fourier transformation is then performed to express the problem in the frequency domain. A root-mean-square approach is used to find the time-averaged power dissipated in tissue corresponding to each frequency. Finally, scaling the electrical conductivity and boundary conditions allows for the solution of the equations at a single frequency, which is used to compute the total frequency-dependent power dissipation at all frequencies.

2.1.1 Problem Definition. Let Ω be an open bounded domain representing the tissue with boundary Γ as in Fig. 2. The time-dependent current density distribution in the tissue obtained from Ampere's law at a material point $\mathbf{x} \in \mathbb{R}^3$ at a time $t \in \mathbb{R}^+$ is given by [28]

$$\nabla \cdot \left(\mathbf{J}(\mathbf{x}, t) + \frac{\partial \mathbf{D}(\mathbf{x}, t)}{\partial t} \right) = 0 \quad \text{on } \Omega \times]0, \tau[\quad (1)$$

where $\mathbf{J}$ (A m^{-2}) is the conduction current density in the tissue and $\mathbf{D}$ (C m^{-2}) is the electric displacement field with the displacement current density given by $\partial \mathbf{D} / \partial t$. $\mathbf{J}$ and $\mathbf{D}$ are related to the electric field $\mathbf{E}$ (V m^{-1}), respectively, by

$$\mathbf{J}(\mathbf{x}, t) = s_e \mathbf{E}(\mathbf{x}, t) \quad (2)$$

$$\mathbf{D}(\mathbf{x}, t) = \epsilon \mathbf{E}(\mathbf{x}, t) \quad (3)$$

where s_e (S m^{-1}) and ϵ ($\text{m}^{-3} \text{kg}^{-1} \text{s}^4 \text{A}^2$) are the scalar electrical conductivity and permittivity of the tissue. The electric field in the tissue is related to the electric potential V by

$$\mathbf{E}(\mathbf{x}, t) = -\nabla V(\mathbf{x}, t) \quad (4)$$

The initial and boundary conditions for the problem in Eq. (1) are given as

$$\left(\mathbf{J}(\Gamma_e, t) + \frac{\partial \mathbf{D}(\Gamma_e, t)}{\partial t} \right) \cdot \mathbf{n} = J_0(t) \quad \text{on } \Gamma_e \quad (5)$$

$$\left(\mathbf{J}(\Gamma_0, t) + \frac{\partial \mathbf{D}(\Gamma_0, t)}{\partial t} \right) \cdot \mathbf{n} = 0 \quad \text{on } \Gamma_0 \quad (6)$$

$$V(\Gamma_g, t) = 0 \quad \text{on } \Gamma_g \quad (7)$$

$$\left(\mathbf{J}(\mathbf{x}, 0) + \frac{\partial \mathbf{D}(\mathbf{x}, 0)}{\partial t} \right) \cdot \mathbf{n} = 0 \quad \text{at } t = 0 \quad (8)$$

with $\Gamma = \Gamma_e \cup \Gamma_g \cup \Gamma_0$, $\Gamma_e \cap \Gamma_g = \emptyset$, $\Gamma_e \cap \Gamma_0 = \emptyset$, and $\Gamma_g \cap \Gamma_0 = \emptyset$. Here, J_0 is the total current density, periodic with a period T (s), $\mathbf{n}$ is the unit outward normal to Γ_e , and V is the electric potential (V).

The power P (W m^{-3}) dissipated in the tissue is then given by

$$P(\mathbf{x}, t) = \mathbf{J}(\mathbf{x}, t) \cdot \mathbf{E}(\mathbf{x}, t) \quad (9)$$

2.1.2 Fourier Transformation. The conduction current density and electric displacement field may be expressed as the following Fourier series:

$$\mathbf{J}(\mathbf{x}, t) = \sum_{n=-\infty}^{\infty} e^{j\omega_n t} \hat{\mathbf{J}}(\mathbf{x}, n) \quad (10)$$

$$\mathbf{D}(\mathbf{x}, t) = \sum_{n=-\infty}^{\infty} e^{j\omega_n t} \hat{\mathbf{D}}(\mathbf{x}, n) \quad (11)$$

where $\hat{\mathbf{J}}(\mathbf{x}, n)$ and $\hat{\mathbf{D}}(\mathbf{x}, n)$ are the Fourier transforms of $\mathbf{J}$ and $\mathbf{D}$, respectively

$$\hat{\mathbf{J}}(\mathbf{x}, n) = \frac{1}{\theta} \int_0^\theta \mathbf{J}(\mathbf{x}, t) e^{-j\omega_n t} dt \quad (12)$$

$$\hat{\mathbf{D}}(\mathbf{x}, n) = \frac{1}{\theta} \int_0^\theta \mathbf{D}(\mathbf{x}, t) e^{-j\omega_n t} dt \quad (13)$$

with $\theta \in \mathbb{N}$. Here, θ is the time period which is related to the angular frequency ω_n (rad s^{-1}) through the following relationship:

$$\omega_n = \frac{2\pi n}{\theta} \quad (14)$$

Replacing Eqs. (10) and (11) into Eq. (1), we can write

$$\nabla \cdot \left(\sum_{n=-\infty}^{\infty} e^{j\omega_n t} \left(\hat{\mathbf{J}}(\mathbf{x}, n) + j\omega_n \hat{\mathbf{D}}(\mathbf{x}, n) \right) \right) = 0 \quad (15)$$

$\hat{\mathbf{J}}(\mathbf{x}, n)$ and $\hat{\mathbf{D}}(\mathbf{x}, n)$ are related to the Fourier transforms of the corresponding frequency domain electric field $\hat{\mathbf{E}}(\mathbf{x}, n)$ as

$$\hat{\mathbf{J}}(\mathbf{x}, n) = s_{e,n} \hat{\mathbf{E}}(\mathbf{x}, n) \quad (16)$$

$$\hat{\mathbf{D}}(\mathbf{x}, n) = \epsilon_n \hat{\mathbf{E}}(\mathbf{x}, n) \quad (17)$$

where $s_{e,n}$ and ϵ_n are the electrical conductivity and the permittivity of the tissue [13], respectively, at frequency ω_n . The permittivity is related to the electrical conductivity by $\omega_n \epsilon_n = s_{e,n} / \tan(\delta_n)$ where δ_n is the loss angle.

Substituting Eqs. (16) and (17) into Eq. (15), multiplying by $e^{-j2\pi n t / \theta}$, and integrating over a period θ we obtain

$$\frac{1}{\theta} \int_0^\theta \sum_{n=-\infty}^{\infty} e^{-j2\pi n t / \theta} e^{j2\pi n t / \theta} \nabla \cdot \left(s_{e,n} \left(1 + j \frac{1}{\tan(\delta_n)} \right) \hat{\mathbf{E}}(\mathbf{x}, n) \right) dt = 0 \quad (18)$$

Knowing that the complex exponentials $e^{j2\pi n t / \theta}$ and $e^{-j2\pi m t / \theta}$ are orthonormal and their inner product is 0 when $n \neq m$, it is shown that each term of the series must satisfy the equality such that

$$\nabla \cdot \left(s_{e,n} \left(1 + j \frac{1}{\tan(\delta_n)} \right) \hat{\mathbf{E}}(\mathbf{x}, n) \right) = 0 \quad (19)$$

For a homogeneous material, assuming that the loss tangent is dependent only on frequency, Eq. (19) simplifies to

$$\nabla \cdot (s_{e,n} \hat{\mathbf{E}}(\mathbf{x}, n)) = 0 \quad \text{in } \Omega, \quad n \in \mathbb{N} \quad (20)$$

Similarly, applying the same transformation to the boundary conditions in Eqs. (5)–(7), we obtain

$$\left(s_{e,n} \left(1 + \mathbf{j} \frac{1}{\tan(\delta_n)} \right) \hat{\mathbf{E}}(\Gamma_e, n) \right) \cdot \mathbf{n} = \hat{\mathbf{J}}_0(n) \quad \text{on } \Gamma_e \quad (21)$$

$$\left(s_{e,n} \left(1 + \mathbf{j} \frac{1}{\tan(\delta_n)} \right) \hat{\mathbf{E}}(\Gamma_0, n) \right) \cdot \mathbf{n} = 0 \quad \text{on } \Gamma_0 \quad (22)$$

$$\hat{V}(\Gamma_g, n) = 0 \quad \text{on } \Gamma_g \quad (23)$$

where $\hat{\mathbf{J}}_0$ and $\hat{V}$ are the Fourier transforms of the total applied current density $\mathbf{J}_0$ at the boundary and the voltage V , respectively.

To obtain the frequency spectrum of the measured input current, the single-sided amplitude spectrum is given by

$$c_n(f) = \begin{cases} \hat{I}_0 & n = 0 \\ 2\hat{I}_0 & n = \{1, 2, \dots, +\infty\} \end{cases} \quad (24)$$

where $\hat{I}_0$ is the Fourier transform of the total current applied at the boundary $\mathbf{I}_0 = A \mathbf{J}_0$, where A is the contact area (m^2) between the electrode and the tissue, and $c_n(f)$ corresponds to an angular frequency ω_n and analog frequency f (Hz) = $2\pi/\omega_n$.

The power dissipated in the tissue is given by

$$P(\mathbf{x}, t) = \sum_{n=-\infty}^{\infty} e^{j\omega_n t} \hat{\mathbf{J}}(\mathbf{x}, n) \sum_{m=-\infty}^{\infty} e^{j\omega_m t} \hat{\mathbf{E}}(\mathbf{x}, m) \quad (25)$$

The time-averaged power P_{avg} over a period θ that is dissipated in the tissue is given by

$$P_{\text{avg}} = \frac{1}{\theta} \int_0^\theta P dt = \frac{1}{\theta} \int_0^\theta \sum_{n=-\infty}^{\infty} e^{j\omega_n t} \hat{\mathbf{J}}(\mathbf{x}, n) \sum_{m=-\infty}^{\infty} e^{j\omega_m t} \hat{\mathbf{E}}(\mathbf{x}, m) dt \quad (26)$$

Applying orthonormality as before where the only nonzero terms are for $m = -n$, we can write

$$P_{\text{avg}} = \sum_{n=-\infty}^{\infty} \frac{1}{\theta} \int_0^\theta \hat{\mathbf{J}}(\mathbf{x}, n) \hat{\mathbf{E}}(\mathbf{x}, n) dt \quad (27)$$

where for a harmonic field, P_{avg} is then

$$P_{\text{avg}} = \sum_{n=-\infty}^{\infty} \frac{1}{2} \text{Re} [\hat{\mathbf{J}}(\mathbf{x}, n) \hat{\mathbf{E}}^*(\mathbf{x}, n)] \quad (28)$$

Using the relationship between current density and electric field in Eq. (16) gives

$$P_{\text{avg}} = \sum_{n=-\infty}^{\infty} \frac{1}{2} s_{e,n} |\hat{\mathbf{E}}(\mathbf{x}, n)|^2 = \sum_{n=-\infty}^{\infty} P_{\text{avg},n} \quad (29)$$

where $P_{\text{avg},n}$ is the time-averaged power dissipated in the tissue associated with a frequency ω_n .

2.1.3 Root-Mean-Square. Time averaging over a period θ , the root-mean-square (RMS) of the electric potential which is sinusoidal at a given frequency is given by

$$V_{n,\text{RMS}}(\mathbf{x}) = \frac{\hat{V}(\mathbf{x}, n)}{\sqrt{2}} \quad (30)$$

and for the $n = 0$ term, the RMS voltage is given by

$$V_{0,\text{RMS}}(\mathbf{x}) = \hat{V}(\mathbf{x}, 0) \quad (31)$$

The RMS of the electric field and the conduction current density are then expressed, respectively as

$$\mathbf{E}_{n,\text{RMS}}(\mathbf{x}) = -\nabla V_{n,\text{RMS}}(\mathbf{x}) \quad (32)$$

$$\mathbf{J}_{n,\text{RMS}}(\mathbf{x}) = -s_{e,n} \nabla V_{n,\text{RMS}}(\mathbf{x}) \quad (33)$$

The RMS of Eq. (20) is

$$\nabla \cdot (s_{e,n} \nabla V_{n,\text{RMS}}(\mathbf{x})) = 0 \quad \text{in } \Omega \quad (34)$$

To compute the RMS of the boundary conditions in Eq. (21), we multiply the left-hand side by its conjugate, integrate over a period T , and take the square root to obtain

$$\begin{aligned} & \left(\frac{1}{\theta} \int_0^\theta s_{e,n}^2 \left(1 + \frac{1}{\tan^2(\delta_n)} \right) \hat{\mathbf{E}}(\Gamma_e, n) \hat{\mathbf{E}}^*(\Gamma_e, n) dt \right)^{1/2} \\ &= \frac{s_{e,n}}{\sin(\delta_n)} \mathbf{E}_{n,\text{RMS}}(\Gamma_e) \end{aligned} \quad (35)$$

The boundary conditions in Eqs. (21)–(23) can then be expressed as

$$s_{e,n} \mathbf{E}_{n,\text{RMS}}(\Gamma_e) \cdot \mathbf{n} = \cos(\varphi_n) \mathbf{J}_{0n,\text{RMS}}(\Gamma_e) \quad \text{on } \Gamma_e \quad (36)$$

$$\mathbf{E}_{n,\text{RMS}}(\Gamma_0) \cdot \mathbf{n} = 0 \quad \text{on } \Gamma_0 \quad (37)$$

$$V_{n,\text{RMS}}(\Gamma_g, t) = 0 \quad \text{on } \Gamma_g \quad (38)$$

where $\mathbf{J}_{0n,\text{RMS}}$ is the RMS of the component of the current boundary condition corresponding to frequency ω_n . The phase angle φ_n is related to the loss angle by $\varphi_n = (\pi/2) - \delta_n$.

The RMS power dissipated in the tissue corresponding to frequency ω_n is then

$$P_{n,\text{RMS}}(\mathbf{x}) = s_{e,0} \mathbf{E}_{n,\text{RMS}}(\mathbf{x}) \cdot \mathbf{E}_{n,\text{RMS}}(\mathbf{x}) \quad (39)$$

2.1.4 Scaled Problem. The problem described in Eq. (32) requires solving the electrical problem at each frequency to obtain the total power dissipated. However, scaling the frequency and the applied current density boundary conditions allows the solution of a single scaled electrical problem instead to find the total power. The frequency dependence of electrical conductivity is shown in Fig. 3(a) where measurements for liver tissue from literature taken at body temperature in the frequency range of 10 Hz–3 MHz are summarized [13–16], while the temperature dependence of the electrical conductivity at different frequencies [17,18] is shown in Fig. 3(b). The reported electrical conductivity of liver can vary by up to 30% in the literature at given frequency and temperature for liver from the same animal species. This may be due to naturally occurring differences in biological tissue that are affected by age, weight, diet, and other physical characteristics [13,17].

The variation of the electrical conductivity is approximately linear with temperature up to 80 °C and shifts upward with increasing frequency, doubling from 100 kHz to 1 MHz. Thus, to capture the behavior of the electrical conductivity with both frequency and temperature, we assume that the temperature-dependent electrical conductivity $s_{e,n}$ at each frequency can be scaled such that

$$s_{e,n}(\omega_n, T) = K_{sn}(\omega_n) s_{e,0}(T) \quad (40)$$

where $s_{e,0}$ is the temperature-dependent electrical conductivity at 400 kHz, and T is the temperature (°C) and K_{sn} is a ratio that depends only on the frequency and is given by

Fig. 3 Variation of electrical conductivity of liver tissue (a) with frequency at body temperature and (b) with temperature at different frequencies

$$K_{sn}(\omega_n) = \frac{s_{e,n}(\omega_n, 37^\circ\text{C})}{s_{e,0}(37^\circ\text{C})} \quad (41)$$

The scaled frequency-dependent electrical conductivity function, K_{sn} , is determined only once using the experimental data set from Refs. [13], [14], and [17] as a property of the tissue and used for all power settings and modes. The variation of the electrical conductivity $s_{e,n}$ with frequency and temperature is shown in a surface plot in Fig. 4.

The applied boundary condition in Eq. (36) can also be scaled by a second scaling factor, K_{jn} , that is constant at a given frequency ω_n and which would be determined from the experimentally measured current density for each power setting and waveform such that

$$J_{0n,\text{RMS}} = K_{jn} J_0 \quad (42)$$

Substituting the scaling factors into the boundary condition in Eq. (9), we can rewrite it as

$$s_{e,0} \nabla V_{n,\text{RMS}}(\Gamma_e) \cdot \mathbf{n} = K_n J_0(\Gamma_e) \quad (43)$$

where J_0 is the scaled current density magnitude, K_n is

$$K_n = \frac{\cos(\varphi_n) K_{jn}}{K_{sn}} \quad (44)$$

The scaled problem is then be stated as

$$\nabla \cdot (s_{e,0}(T) \nabla V_0(\mathbf{x})) = 0 \quad \text{in } \Omega \quad (45)$$

where V_0 is the scaled electric potential in the tissue subject to boundary conditions

$$-s_{e,0}(T) \nabla V_0(\Gamma_e) \cdot \mathbf{n} = J_0(\Gamma_e) \quad \text{on } \Gamma_e \quad (46)$$

$$-s_{e,0}(T) \nabla V_0(\Gamma_0) \cdot \mathbf{n} = 0 \quad \text{on } \Gamma_0 \quad (47)$$

$$V_0(\Gamma_g) = 0 \quad \text{on } \Gamma_g \quad (48)$$

By solving the scaled problem in Eqs. (45)–(48), we obtain the solution for the scaled electric potential in the tissue, V_0 . Therefore, we can then obtain the solution to Eq. (34) as

$$V_{n,\text{RMS}}(\mathbf{x}, t) = K_n V_0(\mathbf{x}) \quad (49)$$

Fig. 4 Variation of the scaled model of electrical conductivity $s_{e,n}$ with frequency and temperature

The corresponding power dissipation $P_{n,\text{RMS}}$ (W) is

$$P_{n,\text{RMS}}(\mathbf{x}) = s_{e,n} \|\nabla V_{n,\text{RMS}}(\mathbf{x})\|^2 = K_{sn} s_{e,0}(T) K_n^2 \mathbf{E}_0(\mathbf{x}) \cdot \mathbf{E}_0(\mathbf{x}) \quad (50)$$

where $\mathbf{E}_0$ is the electric field (V/m) corresponding to V_0 , and the total power applied to the tissue is

$$\begin{aligned} P_{t,\text{RMS}}(\mathbf{x}, t) &= \sum_{n=-\infty}^{\infty} P_{n,\text{RMS}}(\mathbf{x}, t) \\ &= s_{e,0}(T) \mathbf{E}_0(\mathbf{x}) \cdot \mathbf{E}_0(\mathbf{x}) \sum_{n=-\infty}^{\infty} K_{sn} K_n^2 \end{aligned} \quad (51)$$

where we define a power factor F_P as

$$F_P = \sum_{n=-\infty}^{\infty} K_{sn} K_n^2 \quad (52)$$

Table 1 Summary of electrical equations in time domain, frequency domain, and the scaled time-averaged RMS problems

Time domain	Frequency domain	Scaled RMS
$\nabla \cdot \left(\mathbf{J}(\mathbf{x}, t) + \frac{\partial \mathbf{D}(\mathbf{x}, t)}{\partial t} \right) = 0 \quad \text{in } \Omega \times]0, \tau[$ Initial and boundary conditions $\left(\mathbf{J}(\Gamma_e, t) + \frac{\partial \mathbf{D}(\Gamma_e, t)}{\partial t} \right) \cdot \mathbf{n} = \mathbf{J}_0(t) \quad \text{on } \Gamma_e$ $\left(\mathbf{J}(\Gamma_0, t) + \frac{\partial \mathbf{D}(\Gamma_0, t)}{\partial t} \right) \cdot \mathbf{n} = 0 \quad \text{on } \Gamma_0$ $V(\Gamma_g, t) = 0 \quad \text{on } \Gamma_g$ $\left(\mathbf{J}(\mathbf{x}, 0) + \frac{\partial \mathbf{D}(\mathbf{x}, 0)}{\partial t} \right) \cdot \mathbf{n} = 0 \quad \text{at } t = 0$ Power dissipated in the tissue $P(\mathbf{x}, t) = \mathbf{J}(\mathbf{x}, t) \cdot \mathbf{E}(\mathbf{x}, t)$	$\nabla \cdot \left(s_{e,n} \left(1 + \mathbf{j} \frac{1}{\tan(\delta_n)} \right) \hat{\mathbf{E}}(\mathbf{x}, n) \right) = 0 \quad \text{in } \Omega, n \in \mathbb{N}$ $\left(s_{e,n} \left(1 + \mathbf{j} \frac{1}{\tan(\delta_n)} \right) \hat{\mathbf{E}}(\Gamma_e, n) \right) \cdot \mathbf{n} = \hat{\mathbf{J}}_0(n) \quad \text{on } \Gamma_e$ $\left(s_{e,n} \left(1 + \mathbf{j} \frac{1}{\tan(\delta_n)} \right) \hat{\mathbf{E}}(\Gamma_0, n) \right) \cdot \mathbf{n} = 0 \quad \text{on } \Gamma_0$ $\hat{V}(\Gamma_g, n) = 0 \quad \text{on } \Gamma_g$ $P(\mathbf{x}, t) = \sum_{n=-\infty}^{\infty} e^{j\omega_n t} \hat{\mathbf{J}}(\mathbf{x}, n) \sum_{m=-\infty}^{\infty} e^{j\omega_m t} \hat{\mathbf{E}}(\mathbf{x}, m)$	$\nabla \cdot (s_{e,0}(T) \mathbf{E}_0(\mathbf{x})) = 0 \quad \text{in } \Omega$ $-s_{e,0}(T) \mathbf{E}_0(\Gamma_e) \cdot \mathbf{n} = \mathbf{J}_0(\Gamma_e) \quad \text{on } \Gamma_e$ $-s_{e,0}(T) \nabla \mathbf{E}_0(\Gamma_0) \cdot \mathbf{n} = 0 \quad \text{on } \Gamma_0$ $V_0(\Gamma_g) = 0 \quad \text{on } \Gamma_g$ $P_{t,RMS}(\mathbf{x}, t) = F_P s_{e,0}(T) \mathbf{E}_0(\mathbf{x}) \cdot \mathbf{E}_0(\mathbf{x})$

The equations of the electrical problem in the time domain, the transformation to the frequency domain, and then to the time-averaged scaled problem are summarized in Table 1.

2.2 Mixture Theory-Based Model. Electrosurgical heating in soft tissue is modeled using a mixture theory-based thermomechanical model [20]. In this section, we summarize the mass, momentum, and energy balance equations corresponding to this model. The tissue represented by an open bounded domain Ω with boundary Γ , is assumed to be a multiphase mixture consisting of solid (*s*), liquid (*l*), and gas (*g*) phases that occupy the same material point $\mathbf{x} \in \mathbb{R}^3$ at time $t \in \mathbb{R}^+$. The partial density $\hat{\rho}_\alpha$ is defined as the mass of the phase α per unit volume of the mixture at a point $(\mathbf{x}, t)$. The mass balance for the vapor and liquid phase is given by

$$\frac{\partial \hat{\rho}_l(\mathbf{x}, t)}{\partial t} = -\pi_{\text{evap}}(\mathbf{x}, t) \quad \text{in } \Omega \times]0, \tau[\quad (53)$$

$$\frac{\partial \hat{\rho}_g(\mathbf{x}, t)}{\partial t} + \nabla \cdot (\hat{\rho}_g(\mathbf{x}, t) \mathbf{v}_g(\mathbf{x}, t)) = \pi_{\text{evap}}(\mathbf{x}, t) \quad \text{in } \Omega \times]0, \tau[\quad (54)$$

where $\mathbf{v}_g$ is the velocity of the gas phase (m s^{-1}), which is obtained from the linear momentum balance of the gas phase given by Darcy's law

$$\mathbf{v}_g(\mathbf{x}, t) = -\frac{\kappa_g}{\mu_g} \nabla P(\mathbf{x}, t) \quad \text{in } \Omega \times]0, \tau[\quad (55)$$

where κ_g is the permeability of the tissue to the gas phase (m^2), μ_g is the dynamic viscosity of the gas phase ($\text{Pa}\cdot\text{s}$), and P (Pa) is the total hydrostatic pressure which is the intrinsic pressure of the vapor and liquid phases in the tissue. The velocity of the liquid phase was found to be negligible. The linear momentum balance of the solid phase assuming isotropy and linear elasticity is given by

$$\nabla \cdot \boldsymbol{\sigma}_s(\mathbf{x}, t) = \mathbf{0} \quad \text{in } \Omega \times]0, \tau[\quad (56)$$

where the Cauchy stress tensor $\boldsymbol{\sigma}_s$ which is decomposed into the hydrostatic and deviatoric stress τ_s , $\boldsymbol{\sigma}_s$ and τ_s are given, respectively, by

$$\boldsymbol{\sigma}_s = \lambda_s \text{tr}(\boldsymbol{\epsilon}_s) \mathbf{I} + 2\mu_s \boldsymbol{\epsilon}_s = -\varphi_s P \mathbf{I} + \tau_s; \quad \tau_s = \boldsymbol{\sigma}_s - \frac{1}{3} \text{tr}(\boldsymbol{\sigma}_s) \mathbf{I} \quad (57)$$

where λ_s and μ_s are the Lamé coefficients, and $\boldsymbol{\epsilon}_s$ is the solid strain tensor. The energy balance for the mixture by considering an open system is then given by

$$\begin{aligned} (\rho c_p)_{\text{eff}} \frac{\partial T(\mathbf{x}, t)}{\partial t} &= F_P s_{e,0}(T) \mathbf{E}_0(\mathbf{x}) \cdot \mathbf{E}_0(\mathbf{x}) - \hat{\rho}_g(\mathbf{x}, t) \mathbf{v}_g(\mathbf{x}, t) \\ &\quad \cdot \nabla (c_{p,g}(\mathbf{x}, t) T(\mathbf{x}, t)) + \frac{\partial P(\mathbf{x}, t)}{\partial t} \\ &\quad + \nabla \cdot (k \nabla T(\mathbf{x}, t)) - \lambda \pi_{\text{evap}}(\mathbf{x}, t) \quad \text{in } \Omega \times]0, \tau[\end{aligned} \quad (58)$$

such that

$$(\rho c_p)_{\text{eff}} = \hat{\rho}_s(\mathbf{x}, t) c_{p,s}(\mathbf{x}, t) + \hat{\rho}_l(\mathbf{x}, t) c_{p,l}(\mathbf{x}, t) + \hat{\rho}_g(\mathbf{x}, t) c_{p,g}(\mathbf{x}, t) \quad (59)$$

$$k = \varphi_s k_s + \varphi_l k_l + \varphi_g k_g \quad (60)$$

where c_p is the specific heat capacity ($\text{J kg}^{-1} \text{K}^{-1}$) and k is the thermal conductivity ($\text{W m}^{-1} \text{K}^{-1}$), and λ is the specific enthalpy of evaporation (J kg^{-1}). For a differential volume dv , dv_α is the proportion that is occupied by phase α , and the volume fraction is $\varphi_\alpha = dv_\alpha / dv$. The evaporation term π_{evap} is defined explicitly as [29]

$$\pi_{\text{evap}} = \begin{cases} 0 & \text{if } T < 100^\circ\text{C} \\ K(p_{\text{sat}} - P) & \text{if } T \geq 100^\circ\text{C} \end{cases} \quad (61)$$

where K is an empirical evaporation constant, p_{sat} is the saturation vapor pressure at temperature T . The evaporation model is activated once the temperature reaches the boiling point of water (100°C). K is a penalty parameter that must be chosen to be sufficiently large to enforce the evaporation rate to be near its equilibrium value, however, too large a value of K prevents the solution from converging. Hence, based on previous numerical analysis [20], an evaporation constant of $1.5 \times 10^4 \text{ m}\cdot\text{s}$ was used. In addition, to ensure convergence of the solution, a constraint was placed so that the maximum total evaporation losses cannot exceed the total amount of Joule heating applied to the tissue. The power generated in the tissue in the first term on the right-hand side of Eq. (59) is obtained from Eq. (51) by solving for the electric field $\mathbf{E}_0$ in the scaled problem described by Eqs. (45)–(48), and multiplied by the power factor defined in Eq. (52).

2.2.1 Damage Model. During electrosurgery, tissue is thermally damaged when the collagen structure begins to unravel at

higher temperatures [1]. To compare the cumulative damage in the tissue resulting from heating at different power modes, an Arrhenius damage integral [30] is considered such that

$$\Omega(t) = A \int_0^t e^{-\Delta E/RT} dt \quad (62)$$

where $\Omega(t)$ is the cumulative damage integral, A is the frequency factor (s^{-1}), ΔE is the activation energy (J/mol), R is the universal gas constant, and T is the temperature. The factors A and ΔE have been determined for a variety of tissues as they begin to undergo damage, such as skin, collagen, muscular tissue, and liver [30,31], where $A = 7.39 \times 10^{39} s^{-1}$ and $\Delta E = 2.577 \times 10^5 J \cdot mol^{-1}$ for liver tissue.

2.2.2 Initial and Boundary Conditions. For an open bounded domain for the problem with boundary Γ , the part of the boundary exposed to the atmosphere is denoted as Γ_a , and the part of the boundary that is insulated is denoted as Γ_i with $\Gamma = \Gamma_a \cup \Gamma_i$ and $\Gamma_a \cap \Gamma_i = \emptyset$. $\mathbf{n}$ is the unit outward normal to the boundary and $\mathbf{u}_s$ is the displacement of the solid. The initial and boundary conditions for the mass balance equations are given as

Mass Balance:

$$\text{Liquid I.C. } \hat{\rho}_l(\mathbf{x}, 0) = \varphi_{l,0} \rho_l \quad (63)$$

$$\text{Gas I.C. } \hat{\rho}_g(\mathbf{x}, 0) = \varphi_{g,0} \rho_g \quad (64)$$

$$\text{B.C. } \hat{\rho}_g \mathbf{v}_g(\Gamma_a, t) \cdot \mathbf{n} = -\hat{\rho}_g(\Gamma_a, t) \frac{\kappa_g}{\mu_g} \nabla P(\Gamma_a, t) \cdot \mathbf{n} + h_m \varphi_g (\rho_g(\Gamma_a, t) - \rho_{v,amb}) \text{ on } \Gamma_a \quad (65)$$

$$\hat{\rho}_g(\Gamma_i, t) = \varphi_{g,0} \rho_g \text{ on } \Gamma_i \quad (66)$$

where $\varphi_{l,0}$ and $\varphi_{g,0}$ are the initial volume fractions of the liquid and gas phase, respectively. The gas flux at Γ_a is given by the pressure driven flow and mass diffusion to the atmosphere where h_m is the mass transfer coefficient.

The initial and boundary conditions of the momentum balance equations are

Momentum Balance:

$$\text{Solid B.C. } \boldsymbol{\sigma}_s(\Gamma_a, t) \cdot \mathbf{n} = \mathbf{0} \text{ on } \Gamma_a \quad (67)$$

$$\mathbf{u}_s(\Gamma_i, t) = \mathbf{0} \text{ on } \Gamma_i \quad (68)$$

$$\text{Gas I.C. } P(\mathbf{x}, 0) = p_{atm} \quad (69)$$

$$\text{B.C. } P(\Gamma, t) = p_{atm} \text{ on } \Gamma \quad (70)$$

where traction-free boundary conditions are applied to the tissue surface and zero displacement boundary condition on Γ_i . The initial pressure in the tissue and at the boundary Γ is assumed to be atmospheric.

The initial and boundary conditions of the energy balance equations are then given by

Energy:

$$\text{Mixture I.C. } T(\mathbf{x}, 0) = T_0 \quad (71)$$

$$\begin{aligned} \text{B.C. } -k \nabla T(\Gamma_a, t) \cdot \mathbf{n} &= h(T(\Gamma_a, t) - T_{amb}) \\ &+ c_{p,g}(\Gamma_a, t) T(\Gamma_a, t) \left(-\hat{\rho}_g(\Gamma_a, t) \frac{\kappa_g}{\mu_g} \nabla P(\Gamma_a, t) \cdot \mathbf{n} \right. \\ &\left. + h_m (\rho_g(\Gamma_a, t) - \rho_{v,amb}) \right) \text{ on } \Gamma_a \end{aligned} \quad (72)$$

$$\nabla T(\Gamma_i, t) \cdot \mathbf{n} = 0 \text{ on } \Gamma_i \quad (73)$$

where the tissue is initially at a temperature T_0 . The heat flux leaving the tissue on Γ_a is given by the convection and latent losses due to pressure driven flow and diffusion of vapor to the surrounding air.

3 Numerical Results

In this section, we present numerical solutions of the multiphysics model and compare them to in vivo experimental observations on porcine liver tissue. The in vivo experimental approach is summarized in Sec. 3.1., followed by the finite element model and model parameters in Sec. 3.2. Numerical solutions are presented in Sec. 3.3.

3.1 In Vivo Experimental Approach. The in vivo experiments were performed under an approved IACUC protocol at the Albany Medical College, Animal Resources Facility (Albany, NY). Temperature and power measurements were made during electrosurgical heating of liver tissue of five live, anesthetized pigs weighing between 25 and 30 kg with a ball-tipped electrode at different power modes at low- and high-power settings. The abdomen of each pig was sectioned and the liver lobes exposed for the heating procedure. 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. Details of the testing procedure including animal care have been discussed elsewhere [20] and are summarized below.

A stationary electrosurgical heating procedure was performed on the liver tissue using an electrosurgical unit (System 2450® Electrosurgical Generator, Conmed Corporation, Utica, NY) and a 1.17 mm radius bent ball-tip electrode (Electrosurgical Ball Electrode A831, Bovie Medical Corporation, Clearwater, FL). A dispersive return pad (ESU Ground Pad 400-2100, Conmed Corporation, Utica, NY) was placed on the animal's thigh. The electrode was held by a robotic arm (Universal Robots, Setauket, NY) at an angle, and an infrared (IR) camera (FLIR A655sc) with a

Fig. 5 Mesh and boundary conditions of the finite element model of electrosurgical heating of liver

Table 2 List of material parameters used in model [20]

Parameter	Value			Parameter	Value		
Density				Thermal conductivity			
Water	ρ_w	998	kg m^{-3}	Water	k_w	0.57	$\text{W m}^{-1} \text{K}^{-1}$
Vapor	ρ_v	$p/(R_w T)$	kg m^{-3}	Vapor	k_v	0.016	$\text{W m}^{-1} \text{K}^{-1}$
Solid	ρ_s	1520	kg m^{-3}	Solid	k_s	0.28	$\text{W m}^{-1} \text{K}^{-1}$
Specific heat capacity				Convection coefficient	h_c	3.1	$\text{W m}^{-2} \text{K}^{-1}$
Water	$c_{p,w}$	4219	$\text{J kg}^{-1} \text{K}^{-1}$	Mass transfer coefficient	h_m	0.01	m s^{-1}
Vapor	$c_{p,v}$	2026	$\text{J kg}^{-1} \text{K}^{-1}$	Elastic modulus	E	1980	Pa
Solid	$c_{p,s}$	3523	$\text{J kg}^{-1} \text{K}^{-1}$	Poisson's ratio	ν	0.48	
Permeability of gas phase in tissue	κ_g	4.57×10^{-11}	m^2	Initial volume fraction of liquid phase	$\phi_{l,0}$	0.74	
Dynamic viscosity of vapor	μ_g	1.8×10^{-5}	Pa-s	Initial volume fraction of gas phase	$\phi_{g,0}$	$0.1 \phi_{l,0}$	
Enthalpy of evaporation	λ	2.26×10^6	J kg^{-1}

Fig. 6 Voltage and current waveforms measured at 10 W for (a) blend cut, (b) standard coagulation, and (c) spray coagulation power modes. Corresponding Fourier coefficient single-sided amplitude spectrum for 10 W (d) blend cut, (e) standard coagulation, and (f) spray coagulation.

480 pixels $\times$ 640 pixels resolution and 30 Hz data acquisition rate was used to record surface temperature and positioned above the tissue vertical to the tissue's surface.

Three different power modes, "blend cut," "standard coagulation," and "spray coagulation" were applied to the liver at a low-power setting of 10 W and a high setting of 50 W. Each electrosurgical test was repeated twice for each of the power modes and settings on different liver lobes in vivo for each animal. An

oscilloscope (Agilent MSOX2024A) with a 200 MHz bandwidth was used to capture the voltage and current waveform generated by the ESU at a sampling rate of 10 MHz. The current was found by measuring the voltage drop across a 10 Ω resistor placed in series with the tissue, and the resulting waveform amplitude versus time data for both voltage and current was used to determine the modulation period and frequency of the signal. At each measurement, the electrode was placed in contact with the tissue and

Fig. 7 Fourier coefficient single-sided amplitude spectrum at 50W for (a) blend cut, (b) standard coagulation, and (c) spray coagulation

activated for 2 s. The activation time was limited to the time between breaths to avoid errors due to breathing motion. Infrared videos of the temperature field and power measurements were acquired, and prior to testing, an image of a 10 mm $\times$ 10 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. At each infrared image frame corresponding to time instant “t,” temperatures at points with the same radial distance from the tip of the electrode were averaged to compute the radial temperature profile.

3.2 Finite Element Model. The multiphysics model was implemented and solved using COMSOL Multiphysics 5.2 (Comsol Inc., Burlington, MA) and MATLAB R2015a (The MathWorks Inc., Natick, MA). Using a fast Fourier transform in MATLAB, the corresponding Fourier coefficients of the normalized current waveform were determined and the power factor term in Eq. (52) was computed. To elucidate the effect of the frequency dependence of the power generation captured through the multifrequency model, the power factor was also obtained while assuming a single 400 kHz frequency. A 2D axis-symmetric finite element model of the liver tissue with dimensions 30 mm $\times$ 30 mm was developed and discretized using 2162 quadratic quadrilateral finite

elements with four Gauss points per element to capture the large temperature and pressure gradients near the electrode. The ball tip electrode was represented as a rigid body with a tip radius of 1.17 mm. A graded mesh was generated in COMSOL where the mesh was refined near the electrode as shown in Fig. 5 by defining the maximum element size in each section of the geometry. The maximum element size was 2.085×10^{-1} mm near the electrode–tissue interface where sharp temperature gradients occur and a maximum element size of 6.000 mm furthest away where no variation in temperature is observed in the tissue. Prior to heating, the electrode, which is initially at the surface of the tissue, was displaced downwards by 1.8 mm to bring it into contact with the tissue with a finite deformation elasticity model used to solve for the tissue deformation. Figure 5 shows the mesh of the finite element model used with the defined boundary conditions. The material parameters are summarized in Table 2. The finite element simulation was repeated twice, first to capture the temperature variation while including the power factor obtained for the multifrequency model and then to capture the temperature variation while including the power factor of the single-frequency model.

3.3 Simulation Results

3.3.1 Electrical and Frequency Results. The voltage and current signals applied to the tissue in each of the power modes (blend cut, standard coagulation, and spray coagulation) are shown in Figs. 6(a)–6(c) for the 10 W power setting. Using a fast Fourier transform in MATLAB R2015a (The MathWorks Inc., Natick, MA), the corresponding Fourier coefficients of the current waveform normalized by its maximum value, $I_{\text{norm}}(t) = I(t)/I_{\text{max}}(t)$, were determined, and the single-sided amplitude spectrum of the Fourier coefficients $c_n(f)$ are shown for the blend, standard, and spray coagulation power waveforms in Figs. 6(d) and 6(e). The results show that the blend waveform in Fig. 6(a), which is a modulated waveform with a duty cycle of approximately 70%, while not an ideal 400 kHz sine wave, has a dominant frequency of 400 kHz with harmonic frequencies at 800, 1200, and 1600 kHz as in Fig. 6(d) and additional smaller amplitudes at nearby frequencies. The standard coagulation waveform, shown in Fig. 6(b), has a shorter duty cycle of approximately 40%, and at the same power setting, a maximum voltage and current amplitude that is twice that of the blend cut. The waveform shape is nonsinusoidal and the amplitude spectrum of the standard coagulation shows a broad frequency spectrum up to 2.5 MHz. The spray coagulation waveform shown in Fig. 6(c) is similar to that of the standard coagulation but had a short duty cycle (<25%) with a long rest, which means at the same power output, the maximum voltage and current amplitudes are approximately fivefold that of the blend waveform. The amplitude spectrum of the spray coagulation power mode shows a broad frequency spectrum similar to the standard coagulation (4 kHz–2.5 MHz) as in Fig. 6(f). Summarizing the key differences between the three power modes, both coagulation waveforms are nonsinusoidal and with larger peak-to-peak amplitudes and broad spectra with maximum peaks around 220–240 kHz. This is compared with the blend waveform which is dominated by the 400 kHz frequency and is closer to a sinusoidal waveform.

The single-sided amplitude spectrum of the 50 W setting for the three power modes is shown in Figs. 7(a)–7(c). The amplitude spectrum for the blend power mode at 50 W in Fig. 7(a) was found to be similar to that at 10 W with a dominant frequency at 400 kHz and harmonics at 800, 1200, and 1600 kHz. Comparison of the normalized measured voltage and current ($V(t)/V_{\text{max}}$) and ($I(t)/I_{\text{max}}$) at 10 W and 50 W for the blend power mode showed that they are equivalent (<2% error). Therefore, for the blend cut power mode, a normalized general waveform could be used to evaluate the power input to the tissue for all power settings.

The Fourier coefficients for the single-sided frequency spectrum of the standard and spray coagulation mode for the 50 W

Fig. 8 Variation of mean temperature for in vivo electrosurgical heating at 10 W (solid line) with multifrequency (MF) model (dashed line) and single-frequency (SF) model at 400 kHz (dash-dot line) with radial distance for (a) blend cut, (b) standard coagulation, (c) spray coagulation, and with time for (d) blend cut, (e) standard coagulation, and (f) spray coagulation

power settings (Figs. 7(b) and 7(c)) show a significant difference from the 10 W results. This indicates that for the standard and spray coagulation modes, the electrosurgical unit had a significant variation in the waveform shape at different power settings, and thus, the waveform and corresponding frequency spectra at each power setting must be individually considered.

Finally, for both the 10 W and 50 W settings, a small direct current (DC) component was observed in all three frequency spectra of the current that ranged from 0.4 to 35 mA with larger magnitudes at higher power settings and during coagulation, particularly spray coagulation with large peak-to-peak voltage and current amplitudes. Such DC amplitudes can be sufficient to cause muscle

Table 3 Relative L_2 errors between in vivo mean experimental radial and temporal temperature distributions and predicted temperature at 10W power modes by the multifrequency model (MF) and the single-frequency model (SF)

Power mode	t (s)	Radial distribution		Temporal variation		
		MF (%)	SF (%)	R (mm)	MF (%)	SF (%)
Blend cut (10 W)	0.168	1.9	2.9	1.2	1.7	4.7
	0.462	2.4	4.5	1.6	3.8	7.6
	1.008	2.1	5.2	2.0	2.7	5.3
	2.016	2.2	5.7	3.0	1.6	2.1
Standard coagulation (10 W)	0.168	1.7	3.9	1.2	2.3	8.8
	0.462	1.9	5.7	1.6	2.5	9.4
	1.008	1.1	5.9	2.0	1.5	5.8
	2.016	1.9	8.3	3.0	1.3	2.9
Spray coagulation (10 W)	0.168	1.3	3.1	1.2	2.7	9.6
	0.462	2.4	5.6	1.6	2.7	8.6
	0.924	2.3	6.4	2.0	1.4	3.6
	1.890	2.0	6.3	3.0	0.8	1.4

stimulation [32], however, no such stimulation was observed during experiments. This may be due to the electrode being applied to the liver rather than muscular tissue with the current density distribution around the electrode falling off as the inverse of the square of the radial distance from the electrode. However, this DC component may become a concern when multifrequency power modes are applied to muscular tissue such as cardiac tissue or areas with significant enervation [33].

3.3.2 In Vivo Temperature Results. The mean experimental radial temperature distribution for the in vivo electrosurgical heating experiments at 10 W are shown in Figs. 8(a)–8(c) for the blend cut, standard coagulation, and spray coagulation modes at different instances of time. The variation of temperature with time is shown in Figs. 8(d)–8(f) for these modes at different radial distances from the electrode. The mean experimental temperature is compared to the temperature prediction of the multifrequency model and a single-frequency model assuming a nominal frequency of 400 kHz. The prediction accuracy of the two models is compared using the evaluating the relative L_2 error defined as

$$\varepsilon = \sqrt{\frac{\sum_{i=1}^N (T_{\text{exp},i} - T_{\text{sim},i})^2}{\sum_{i=1}^N (T_{\text{exp},i})^2}} \quad (74)$$

where $T_{\text{exp},i}$ and $T_{\text{sim},i}$ are the mean experimental temperature and simulation temperature, respectively, at a location i along the temperature profile, and N is the total number of points in the profile.

For each 10 W power mode, the relative L_2 error for the radial distribution and temporal variation of temperature are summarized in Table 3. The temperature variation with radial distance and time for the 10 W blend cut mode in Figs. 8(a) and 8(d) shows that the multifrequency model is able to closely capture the temperature increase and heating in the tissue within an error range of 1.6–3.8%. In comparison, the results of the single-frequency model show a small under-prediction in temperature particularly near the end of the electrosurgical heating process at 2.016 s and near the electrode at a radial distance of 1.6 mm where the maximum error is 7.6%. This under-prediction is attributed to a smaller heat generation due to neglecting the frequency dependence of the electrical conductivity. Since the blend cut waveform is a modulated 400 kHz sine wave and the amplitude spectrum is dominated by the 400 kHz term, neglecting the smaller frequency contributions across the spectrum and the harmonic frequencies results in a relatively small increase in model prediction error.

For the standard coagulation power mode, which has a broad frequency spectrum, a more significant difference between the multifrequency and single-frequency models is observed. The multifrequency model accurately predicts radial and temporal temperature variation when compared to the mean experimental temperature in Figs. 8(b) and 8(e), with an error range of 1.1–2.5%. In comparison, neglecting the frequency-dependence of the electrical conductivity in the single-frequency model results in a more significant under-prediction of tissue temperature that increases with time due to the cumulative effect of a reduced Joule heating rate in the tissue. This increases the error between the single-frequency model temperature prediction and the mean experimental temperature with increasing time, and the maximum error of 9.4% occurs close to the electrode at a radial distance of 1.6 mm. Similarly, for the spray coagulation mode in Figs. 8(c) and 8(f), the multifrequency model closely captures the experimental temperature variation with radial distance and time with an error range of 0.8–2.7%. The under-prediction of temperature that increases due to neglecting frequency-dependence also occurs in the spray coagulation power mode where the maximum error of 9.6% occurs close to the electrode at a radial distance of 1.2 mm. These results show that in coagulation modes which have a broad frequency spectrum, the multifrequency model is able to capture the increased Joule heating due to the frequency-dependent electrical properties and provides a more accurate temperature prediction when compared to experiment.

The temperature predictions of the multifrequency and single frequency models are compared with the mean experimental radial temperature distribution and temperature variation with time for the 50 W in vivo electrosurgical experiments and shown in Figs. 9(a)–9(f) for the blend cut, standard coagulation and spray coagulation settings, respectively. For each power mode, the relative L_2 error between the measured temperature and model predictions for the radial distribution and temporal variation are summarized in Table 4. For the blend power mode, the comparison of the multifrequency model temperature prediction with experimental temperature in Figs. 9(a) and 9(d) shows no significant improvement over the single-frequency model results. The comparison of the error for the radial and temporal distributions shows a small reduction in error for all profiles in the multifrequency model, with a small increase occurring near the electrode at a radial distance of 1.2 mm. The single-frequency model has a smaller Joule heating rate and therefore results in less error close to the electrode. The increase in error in the multifrequency model near the electrode is attributed to evaporation occurring below 100 °C, which is not sufficiently captured by the surface vapor diffusion or the evaporation model, as well as changes to the thermal contact properties due to tissue desiccation or charring at the electrode surface. In the single-frequency model, the overall heat generation is under-predicted and this may mask the effect of evaporation losses below 100 °C and changes in contact properties on temperature prediction.

In comparison, the standard coagulation results in Figs. 9(b) and 9(e) show a significant difference in the temperature predicted by the multifrequency and single-frequency models when compared to experimental results. The multifrequency model captures the temperature variation with radial distance and time with an L_2 error range of 2.9–8.2%. The maximum error occurs at 0.210 s and close to the electrode at a radial distance of 1.2 mm where, as in the blend mode, the model does not sufficiently capture surface evaporation losses below 100 °C or possible changes in the electrode–tissue contact properties. However, once the temperature reaches 100 °C near the electrode, the evaporation model is activated and further increase in temperature is prevented. This allows the model to catch up to the mean experimental temperature and significantly reduces the maximum error at the final time of 0.462 s. The single-frequency model under-predicts the experimental temperature where the L_2 error in the radial temperature distribution increases with time. For the temperature variation in time, the reduced heat generation in the single-frequency model

Fig. 9 Variation of mean temperature for in vivo electrosurgical heating at 50 W (solid line) with frequency-dependent waveform model (dashed line) and 400 kHz model (dash-dot line) with radial distance for (a) blend cut, (b) standard coagulation, (c) spray coagulation, and with time for (d) blend cut, (e) standard coagulation, and (f) spray coagulation

results in a smaller error at a radial distance of 1.2 mm near the electrode than the multifrequency model. However, further from the electrode where evaporation does not occur, the error is significantly higher and reaches a maximum of 13.0% at a radial distance of 1.6 mm. Similarly, the results of the spray coagulation in Figs. 9(c) and 9(f) show that the multifrequency model predictions

closely capture variation of the experimental temperature with radial distance and time with an L_2 error range of 0.9–5.3% with a larger error of 9.1% occurring near the electrode at 1.2 mm. The single frequency again shows a significant increase in almost all error measures, with a smaller error in the vicinity of the electrode. These results show that for both the standard and spray

Table 4 Relative L_2 errors between in vivo mean experimental temperature distributions and predicted temperature at 50 W power modes by the multifrequency model (MF) and the single-frequency model (SF)

Power mode	Radial distribution			Temporal variation		
	t (s)	MF (%)	SF (%)	R (mm)	MF (%)	SF (%)
Blend cut (50 W)	0.042	1.2	1.2	1.2	8.8	6.1
	0.168	1.4	3.1	1.6	3.1	6.2
	0.210	3.8	4.8	2.0	1.9	3.7
	0.378	5.2	6.3	3.0	0.9	0.9
Standard coagulation (50 W)	0.042	3.0	3.5	1.2	8.2	4.1
	0.126	3.2	7.9	1.6	2.6	13.0
	0.210	5.4	9.0	2.0	5.6	12.8
	0.462	4.1	10.8	3.0	2.9	5.3
Spray coagulation (50 W)	0.042	1.8	2.1	1.2	9.1	5.3
	0.084	1.6	5.4	1.6	2.6	10.8
	0.210	3.6	8.1	2.0	1.7	6.5
	0.420	5.3	10.5	3.0	0.9	2.5

Fig. 10 Damage distribution in the tissue near the electrode after 2 s of heating at 10 W for (a) blend cut, (b) standard coagulation, and (c) spray coagulation

coagulation modes, which have broad frequency spectra, frequency dependence of the electrical conductivity has a significant effect on the heat generation rate that affects the accuracy of temperature predictions.

To compare the difference in total heating between different power modes, the Arrhenius damage predicted by the multifrequency model in the vicinity of the electrode and tissue interface is shown in Figs. 10(a)–10(c) after 2 s of heating at 10 W for the blend cut, standard coagulation, and spray coagulation modes, respectively. The maximum tissue damage was found to occur below the electrode, where the Arrhenius damage integral was 3.06×10^4 for the blend cut, 2.65×10^4 for the standard coagulation, and 2.14×10^4 for the spray coagulation power modes. Therefore, the blend cut mode results in the most damage compared to the standard coagulation and spray coagulation modes where the maximum damage is 13.4% and 30.1% smaller, respectively. These results show that greater damage is correlated with modes having larger duty cycles and a spectrum that peaks at larger frequencies. Since the coagulation modes have broad frequency spectra which peak at frequencies below 400 kHz where the conductivity is smaller, this results in a smaller root-mean-square value of the measured current. Consequently, there is less heat generation and cumulative thermal damage in the tissue in the two coagulation modes when compared to the blend power in the temperature range considered in this study, which is below 105 °C.

4 Conclusions

In conclusion, a model that can capture tissue heating during multifrequency power modes was proposed and compared with mean experimental temperature measured during in vivo electrosurgical heating of porcine liver tissue. The multifrequency model captures the frequency dependence of electrical properties and the corresponding effect on the rate of heat generation in the tissue through a single power factor, allowing for a simple integration with any model of tissue heating. For the power modes considered, coagulation modes showed a frequency spectrum that was broad, while the blend mode was dominated by pure cut frequency. The results showed that frequency dependence of the electrical properties can have a significant effect on the predicted temperature distribution and that neglecting the frequency dependence results in significantly higher errors. Comparison between the damage profiles for the three power modes showed that more damage correlated with larger duty cycles. Further improving the accuracy of the conductivity and contact properties at the electrode–tissue interface along with surface evaporation losses below 100 °C, which could be affected by water content, temperature, and tissue damage, could further enhance solution accuracy near the electrode–tissue interface at higher power settings. Measuring the changes in electrical and thermal tissue properties at temperatures above 100 °C along with a more complete damage model that accounts for thermal, mechanical changes due

to denaturation and shrinkage of the tissue as well as desiccation effects could enhance the understanding of the different power modes and their appropriate use in different surgical procedures.

Funding Data

- National Institutes of Health (Grant Nos. R01EB005807, R01EB009362, R01EB010037, R01EB010037, R01HL119248, and R44OD018334; Funder ID: 10.13039/1000000002).

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