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Influence of temperature-dependent tissue parameters on monopolar coagulation model

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Abstract: The use of high-frequency (HF) generators for HF surgical treatment of biological tissue has become an indispensable part of today's surgical applications of all kinds. Generally, HF alternating electric currents between 300 kHz and 1 MHz are used to induce hemostasis by heating at the cellular level. This effect can be attributed to Joule heating, in which a current dissipated conductor converts the electrical energy into thermal energy. However, sound evidence on the reliability and effectiveness of application-specific HF generator modes is not sufficiently available. Usually, the evidence takes place empirically by means of preclinical or clinical studies. Nevertheless, a corresponding empirical data collection is time- and cost-intensive. Therefore, physiological and statistical modeling provides the opportunity to relate tissue response to the applied electrical energy to obtain a prediction of the tissue reaction.

In this contribution, we establish a monopolar coagulation model of an already well-known model approach based on Pennes bioheat equation. Additionally, the vaporization of tissue water at the water boiling point is considered. Furthermore, a variation of temperature-dependent tissue parameters was performed to analyze their impact on the model output.

The simulation results demonstrate that the initial electrical conductivity has the greatest influence on the temperature distribution as well as on the time until the tissue temperature reached the boiling point of water. In contrast, the tissue water content has the least impact on the model output. Depending on the desired coagulation effect, HF power control as a function of electrical conductivity or its reciprocal, tissue resistance, must be added next in an improved model.

Keywords: monopolar coagulation, HF modeling, FEM simulation, temperature-dependent tissue parameter

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1 Introduction

High-frequency (HF) surgery is a commonly used surgical procedure and can be utilized in many different applications. Generally, it is used to achieve hemostasis by means of HF electric current. A sufficiently reliable modeling approach for HF current-induced heating can predict tissue behavior to provide evidence for the functionality of a system device in use or can be used for developing purposes, which can save time and money for clinical approval.

Various mathematical approaches have already been developed and investigated by many other research groups. Mainly, their research aspects were focused on radiofrequency ablation (RFA) for tumor treatment [1] or bipolar application method vessel sealing [2]. However, these two specific HF techniques have limited applicability when modeling a standard HF application method, such as monopolar coagulation with a ball electrode in the here considered case.

An often-used model is the bioheat transfer model or the so-called Pennes Bioheat equation [3]. Over the last years, many researchers investigated Pennes bioheat equation to further optimize it [2, 4] and compared it to other mathematical approaches [5]. All in all, the bioheat equation has proven to be a good approximation for heat distribution in biological tissue. Therefore, this model is also used in this contribution to analyze temperature distribution in a monopolar coagulation process with variations of initial values for electrical and thermal conductivity. Additionally, the influence of the initial tissue water content was analyzed.

2 Materials and methods

A finite element (FE) model was established with a 4 mm ball electrode in contact with biological liver tissue. The model represents a monopolar coagulation process. The temperature distribution in the tissue was simulated. For the simulations of the distribution of the applied electrical field as well as the resulting heat distribution, we used COMSOL Multiphysics 6.0 (Burlington, MA). The simulations were performed on a workstation with AMD Ryzen 7 3700x 8-Core processor (3.59 GHz) and 64 GB RAM.

2.1 Mathematical model

When modeling heat distribution in biological liver tissue induced through HF surgical applications, a partial differential equation (PDE), the so-called bioheat equation [3] can be used to represent the thermally and electrically coupled problem. This equation is applied to govern the heat distribution in the biological tissue

$$\rho c_{\text{eff}}(T)\frac{\partial T}{\partial t} = \nabla \cdot k(T)\nabla T + \boldsymbol{J}\cdot\boldsymbol{E} - \omega_b c_b \rho_b (T - T_b) + Q_m, \quad (1)$$

where ρ represents the tissue density (1079 kg/m³) [7], $c_{\text{eff}}(T)$ is the temperature-dependent effective heat capacity (J/(kg·K)), T is the temperature (°C), t is the time (s), $k(T)$ is the temperature-dependent thermal conductivity (W/(m·K)), $\boldsymbol{J}$ and $\boldsymbol{E}$ are the current density (A/m²) and the electrical field intensity (V/m) respectively, ω_b is the effective capillary blood perfusion parameter (0.0064 1/s) [6], c_b is the blood heat capacity (3617 J/(kg·K)) [7], ρ_b is the blood density (1050 kg/m³) [7], T_b is the capillary blood temperature (37 °C) and Q_m is the metabolic heat generation rate (W/m³) of the tissue. Since the applied electrical energy and therefore, the dissipation due to Joule heating $Q_{el} = \boldsymbol{J}\cdot\boldsymbol{E}$ in this model is orders larger than the impact of the metabolic heat generation rate, Q_m can be neglected, i.e., $Q_m = 0$. Furthermore, the electrical field intensity is defined by $\boldsymbol{E} = -\nabla V$ and the current density by Ohm's law with $\boldsymbol{J} = \sigma(T)\boldsymbol{E}$.

The electrical field is determined by a quasi-static electrical approach using the Laplace equation to obtain the electrical field in the ball electrode and the tissue

$$\nabla \cdot \sigma(T)\nabla V = 0, \quad (2)$$

where $\sigma(T)$ is temperature-dependent electrical conductivity (S/m) and V is the voltage (V).

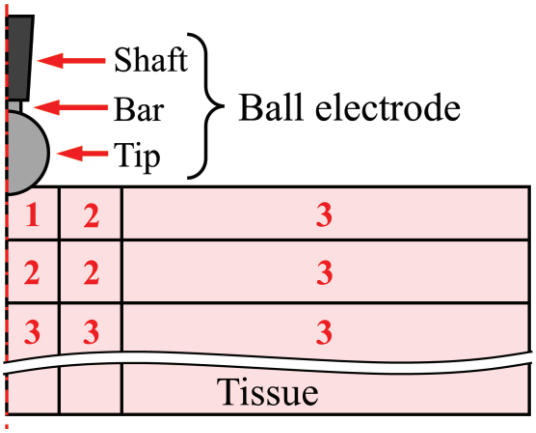

Figure 1: 2D axisymmetric schematic of monopolar coagulation with 4 mm ball electrode. The tissue was separated into three different regions (1, 2, and 3) depending on the distance to the touching electrode tip.

2.2 Material properties of the model

Properly defined material properties especially temperature-sensitive parameters are essential for reliable results of the simulation. However, it is not necessary to model all material properties, e.g., as a function of the temperature. Therefore, the material properties of stainless steel for the electrode tip ($\sigma_{\text{tip}} = 1.4\ \text{MSm}^{-1}$, $k_{\text{tip}} = 25\ \text{W}(\text{m}\cdot\text{K})^{-1}$, $\rho_{\text{tip}} = 7.7\ \text{kg/dm}^3$ and $c_{\text{p,tip}} = 460\ \text{J}(\text{kg}\cdot\text{K})^{-1}$) and the electrode shaft are considered to be constant. Conversely, tissue properties like the thermal and electrical conductivity as well as the specific heat can have an impact on the simulated results when their temperature dependence is not taken into account. In addition, the water content in the tissue plays a crucial role in electrical conductivity and specific heat. Heating the tissue leads at a certain temperature to evaporation of tissue water and, thus, desiccation of tissue. This water loss changes the tissue properties. Therefore, the approaches from Yang et al. [4] and Chen et al. [2] are taken into account.

Both the thermal and the electrical conductivity were considered as function of the temperature. For the thermal conductivity, we used $k(T) = k_{\text{ref}} + k_1(T - T_{\text{ref}})$, where k_{ref} is the thermal conductivity at $T_{\text{ref}} = 25°\text{C}$ and k_1 is the coefficient (0.001161 W/(m·K·°C)) [8] to control the increase of k according to the temperature. For the electrical conductivity, an increase of 2 %/°C [2] was used for temperatures below 100 °C. Accordingly, $\sigma(T) = \sigma_{\text{ref}}[1 + 0.02(T - T_{\text{ref}})]$ when $T < 100$ °C where σ_{ref} is the electrical conductivity at $T_{\text{ref}} = 25$ °C. Due to evaporation of tissue water, an electrical conductivity of 0.01 S/m, following Chen et al. [2], was applied for $T \geq 100$ °C. For a smooth transition at the boiling point of 100 °C in the function $\sigma(T)$, the continuous first derivative was selected as smoothing option in COMSOL. To consider the effect of tissue desiccation, the amount of tissue water can be described by a logistic function in the style of Chen et al. [2]. Therefore, the tissue water content is given by $W(T) = W_{\text{ref}}/\left(1 + e^{\beta(T - T_s)}\right) + W_{\text{end}}$ where W_{ref} is the water amount at $T_{\text{ref}} = 25\ °C$, β is the coefficient (0.35) to control the steepness of water loss, T_s is the boiling temperature threshold (100 °C) and W_{end} is the assumend remaining water amount (0.001 %) after the coagulation process. Due to the dependence of the electical conductivity from the tissue water content, we multiplied $\sigma(T)$ by the relative water amount.

The effective specific heat of tissue was modeled regarding Chen et al. and consists of the specific heat of liver tissue (c_t = 3540 J/(kg·K)) [7], water (c_w = 4178 J/(kg·K)) [7], and latent heat (c_l = 2.26e6 J/kg) through water loss [2]. Thereby c_w and c_l depend on $W(T)$ and the first derivative of $W(T)$, respectively. All properties not mentioned so far and used in the model are listed in Table 1.

Table 1: List of material properties used in the FE model and for the parametric sweeps.

Parameter	Table head 2	Source
Thermal conductivity k_{ref}	[0.42 0.52 0.62] [W/(m·K)]	[7]
Electrical conductivity σ_{ref}	[0.13 0.33 0.55] [S/m]	[7]
Water content W_{ref}	[60 70 80] [%]	

2.3 FE model and boundary conditions

The software COMSOL Multiphysics was used to create a 2D axisymmetric FE model of a 4 mm ball electrode in contact with biological tissue. The insertion depth of the electrode into the tissue was 0.2 mm. The electrode was built of three parts, the electrode shaft (only for illustrative purposes), the electrode tip, and an electrode bar (0.11 cm in diameter and a length of 0.2 cm) to connect shaft and tip. The tissue itself was modeled with a side length of 2.5 cm. For a better mesh control in the tissue region near the contact area to the ball electrode, we added two layers on the left and the top side of the tissue. The first layer has a thickness of 0.25 cm and the second one 0.3 cm. Through the added two layers, the tissue has nine partitions which were grouped into three different regions 1, 2, and 3 (see Figure 1).

The automatic meshing generator of COMSOL Multiphysics was used for triangular meshing of the build geometries. However, we predefined different mesh sizes for the tissue regions 1 to 3. For region 1 being in contact with the electrode, a maximum element mesh size of 0.01 mm was chosen. The maximum element size of region 2 and 3 was defined as 0.02 mm and 0.0345 mm, respectively. Furthermore, a fine mesh size was predefined for the electrode tip, bar, and shaft. To prevent the model from running into a singularity during the calculation, a corner refinement for the electrode tip and tissue region 1 with an element size scaling factor of 0.1 was used. The model consisted of 17001 elements in total.

This FE model is the basic prerequisite for solving the already described thermally (see eq. (1)) and electrically (see eq. (2)) coupled problem of HF induced tissue heating. Nevertheless, well-set initial conditions and boundary conditions are required. For the electrical problem, the ball electrode (65 V), as well as the neutral electrode (the bottom and right side tissue boundary (0 V)), were defined as a Dirichlet boundary condition. The top side of the tissue, which is not in contact with the ball electrode, the passage from electrode bar to electrode shaft, and the outer boundary of the ball electrode, which has no contact with the tissue was defined as Neumann boundary conditions, i.e., $\boldsymbol{J} = 0\,\mathrm{A/m^2}$.

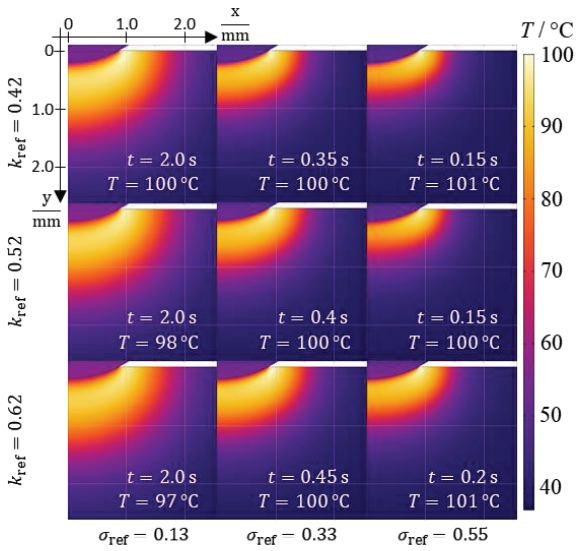

Figure 2: Temperature distribution in biological tissue with three different initial electrical σ_{ref} (S/m) and thermal k_{ref} (W/(m·K)) conductivity values when either of the limits is reached, (1) tissue temperature reached the boiling point of water, or (2) end of simulation time reached without reaching the boiling point. Thereby, the electrical and the thermal conductivity is increasing from left to right and from top to bottom, respectively. W_{ref} was 80 %.

For the initial condition, we set the electric potential for all relevant surfaces to zero.

For the bioheat transfer, an initial condition of 37 °C was set for all tissue regions and the ball electrode. The tissue surface on top that has no ball electrode contact and the ball electrode surface, which has no contact with the tissue was defined to have no heat flux (Neumann boundary condition). The other tissue surfaces were defined to stay at a constant temperature of 37 °C.

The linear direct solver PARDISO from COMSOL Multiphysics was used to solve our model. As HF application time, we assumed 2 seconds with a step size of 0.05 seconds. The parametric sweep of σ_{ref}, k_{ref} and W_{ref} were performed for all possible combinations. The considered values are listed in Table 1. The computation time was 1 h 16 min.

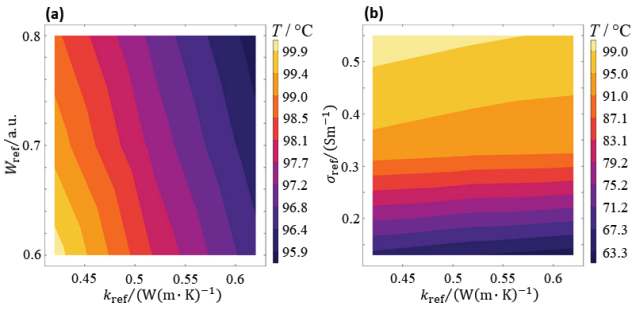

Figure 3: Sensitivity of the maximum output temperature by varying σ_{ref}, k_{ref} and W_{ref} when 100 °C is reached. (a) shows the comparison of W_{ref} over k_{ref} at $\sigma_{\mathrm{ref}} = 0.13\ \mathrm{Sm^{-1}}$ after a simulation time of 1.75 s. (b) shows the comparison of σ_{ref} over k_{ref} at $W_{\mathrm{ref}} = 0.8$ after a simulation time of 0.15 s.

3 Results and Discussion

Figure 2 shows the simulated temperature distribution in the tissue at different initial values for σ_{ref} and k_{ref} at W_{ref} of 80 %, when either the temperature in the tissue has reached the boiling point of water or the maximum simulation time of 2 s has been reached. The results revealed that the temperature output of the model highly depends on the electrical conductivity of the tissue. This is illustrated by the time t required to reach the boiling point. The time to reach the boiling point is reduced by about 82.5 % when σ_{ref} is initially increased from 0.13 S/m to 0.33 S/m and can be further reduced by more than half from 0.33 S/m to 0.55 S/m. Another indicator of the dependency on σ_{ref} is the propagation depth of the temperature, which we defined as the point at which the tissue still reaches 60 °C. The propagation depth along the symmetry axis is reduced on average by about 45 % from 0.13 S/m to 0.55 S/m. The faster the boiling point in the tissue near the electrode is reached, the least temperature can distribute into deeper tissue regions. In real applications, a variation of σ_{ref} can occur due to mixed tissue, the transition from one tissue to another, or simply due to desiccation of the tissue. This can lead to an increase of tissue resistivity. Since σ_{ref} can thus change significantly, it is important to control the power as a function of the measured tissue resistivity to achieve the desired coagulation effect. This also means that the required coagulation effect needs to be known. For deep coagulation, it is necessary to increase the tissue temperature slowly – less power over a longer period (a few seconds). Superficial coagulation, on the other hand, requires a faster temperature rise – higher power over a short period (a few milliseconds). Thus, the electrical conductivity or its reciprocal value, the resistivity, has to be included in the calculation of the power output of the generator for the desired coagulation effect. In modern HF generators, a control loop is already integrated, but not in this simulation. To consider this in future simulations, we need to improve the model.

Increasing the initial thermal conductivity leads to a better temperature distribution in the tissue and therefore, increases the time until the boiling point is reached by about 30 % from 0.42 W/(m·K) to 0.62 W/(m·K). However, a temperature hotspot near the tissue surface at the edge of the electrode-tissue junction is noticeable in all combinations.

Figure 3a demonstrates that an increase in thermal conductivity leads to a bigger impact on the maximum output temperature than a variation of the initial tissue water content. Comparing the impact of electrical and thermal conductivity, a change in electrical conductivity has a larger effect on the model output (Figure 3b). It also shows that a linear increase in σ_{ref} does not result in a linear increase in temperature above 0.33 S/m.

Finally, the following additional limitations identified for this study included: no tissue compression at the electrode to tissue contact area, use of direct electric current instead of alternating HF current, no tissue shrinkage due to desiccation, no changing contact resistance, and no heat convection to ambient air. These limitations need to be further investigated.

4 Conclusion

In this contribution, we presented a monopolar coagulation FE model with varying temperature-dependent tissue parameters solved by the bioheat equation. The influence of the varying model parameters on the temperature distribution in the tissue was investigated. Thereby, the simulation results revealed that the initial tissue water content has the least and the electrical conductivity has the biggest impact on the tissue temperature.

Author Statement

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