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Dependency on thermal and electrical conductivity in monopolar coagulation

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Abstract: High-frequency surgery is a commonly used surgical procedure to mainly achieve hemostasis. In monopolar coagulation, tissue is heated so that it shrinks. A reliable model of such a procedure can help to understand the physical processes ongoing. Based on a previous investigation, a monopolar coagulation model was used to analyze the global change in current and tissue resistance. Different initial parameter values for thermal and electrical conductivity were used. The results show that a change in electrical conductivity leads to a significant change in the current flow and tissue resistance.

I. Introduction

For decades, high-frequency (HF) alternating current (AC) has been used in surgical applications to heat tissue and thus specifically stop bleeding. The heating of the tissue is due to Joule heating, in which the flow of current through a conductor converts some of the electrical energy into thermal energy. For medical approval of such products, evidence of safety and efficacy is required. This is done based on empirical data, which is, however, time- and cost-intensive to obtain. A validated model can remedy this and support the approval procedure.

In a previous investigation [1], we analyzed the effect of parameter variations on the temperature distribution in biological tissue based on an already-established model. Due to the heating of the biological tissue, temperature-dependent parameters change. Consequently, if a constant DC voltage is applied, the current through the tissue is regulated. Therefore, this contribution analyzes how the global tissue resistance and current flow change during the modeled coagulation process at a constant applied voltage as the initial thermal and electrical conductivities vary.

II. Material and methods

For the simulation of the heat generated by electric current and its temporal and spatial distribution in biological tissue, a finite element (FE) model was established. The FE model represents a monopolar coagulation process of biological tissue using a ball electrode and is described in more detail by Busch et al. [1].

The temporal and spatial distribution of the applied electrical potential as well as the resulting temperature distribution in the biological tissue can be described by a partial differential equation (PDE), the so-called Pennes bioheat equation [2]. For a more realistic model, the temperature dependency of some tissue parameters such as electrical conductivity, thermal conductivity, and effective heat capacity was considered.

For the approximation of the temperature-dependent parameters, a linear function for thermal conductivity and a piecewise function for electrical conductivity was used as described in Busch et al. [1]. The representation of the effective heat capacity was modeled according to Chen et al. [3]. Three different values were used for the initial thermal conductivity k_{ref} and the initial electrical conductivity σ_{ref}, respectively. For k_{ref} we used 0.42 W/(m·K), 0.52 W/(m·K), and 0.62 W/(m·K). For σ_{ref} we used 0.13 S/m, 0.33 S/m, and 0.55 S/m. The initial water content W_{ref} was set to 80 %.

The common quasi-static approach was used to determine the electric field in the tissue by solving Laplace's equation [1]. This approximation is based on the negligibility of displacement currents and the purely resistive behavior of biological tissue for HF voltages. Therefore, a constant direct current (DC) voltage of 65 Volts was applied instead of a pure 350 kHz sinusoidal voltage as used in the considered coagulation application. To ensure that the electrically and thermally coupled problem can be solved, reliable boundary and initial conditions were applied as they are described in Busch et al. [1].

The simulation was performed using the software COMSOL Multiphysics for an application time of two seconds. The resulting global current I flow through the tissue over time t was determined. The global tissue resistance R was derived using Ohm's law. The results were analyzed and plotted using the software MATLAB from MathWorks.

III. Results and discussion

The simulated results of $I(t)$ and $R(t)$ for the six different combinations of k_{ref} and σ_{ref} are shown in the three graphs a) to c) of Fig. 1. Where Fig. 1a) shows the curves at σ_{ref} = 0.13 S/m, Fig. 1b) at σ_{ref} = 0.33 S/m and Fig. 1c) at σ_{ref} = 0.55 S/m. The curves in all three graphs indicate, that the increase of k_{ref} leads only to a minor change in $I(t)$ and

$R(t)$ compared to a change of σ_{ref}. The inverse proportionality of $I(t)$ to $R(t)$ given by Ohm's law is shown clearly in all three graphs.

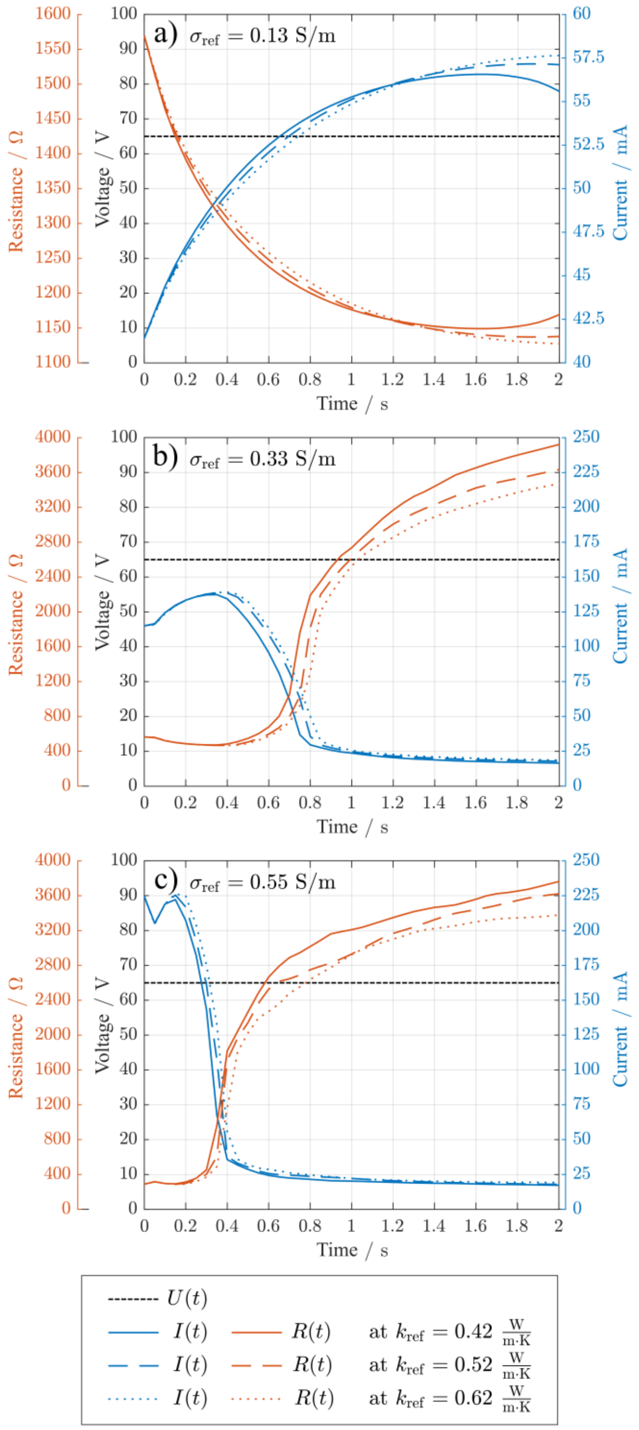

Figure 1: Simulation results of global current $I(t)$ and tissue resistance $R(t)$ at a constant applied DC voltage $U(t)$ of 65 V of a tissue coagulation with three different initial values of thermal conductivity at an initial electrical conductivity of a) σ_{ref} = 0.13 S/m, b) σ_{ref} = 0.33 S/m, and c) σ_{ref} = 0.55 S/m.

When considering only the dotted lines, corresponding to the data given by k_{ref} = 0.62 W/(m·K), an increase of σ_{ref} changes $R(t)$ significantly. This can be seen in Fig. 1a) with σ_{ref} of 0.13 S/m, where $R(0)$ is 1570 Ω and R strictly decreases over the simulated coagulation process by up to 442 Ω. In Fig. 1b) with σ_{ref} of 0.33 S/m $R(0)$ is reduced to 565 Ω and decreases only for the first 0.4 sec by 98 Ω. After a time of 0.4 sec the resistance starts to increase and ends after 2 sec at 3470 Ω. $R(0)$ is even lower at 290 Ω in Fig. 1c) with σ_{ref} of 0.55 S/m and drops by only 3 Ω until the resistance rises again after 0.15 sec.

The decrease in resistance, at the beginning of the coagulation process, is due to the heating of the tissue which leads to an increase in the electrical conductivity. However, this only happens until tissue water starts to decrease through evaporation. The reduction of tissue water results in a decrease in electrical conductivity, which increases the tissue's overall resistance.

Comparing the results from this contribution with the results of our previous study in [1], where we performed the simulation only until the boiling point is reached, we can conclude that the boiling point is reached immediately after the first zero crossing with a negative slope of the first derivative of the current curve.

The used model has some known limitations. First, for simplicity, a constant DC voltage was used instead of an AC voltage as in real HF applications. If an AC voltage is considered in the model, the tissue impedance instead of the resistance needs to be analyzed. Furthermore, these simulations did not consider the tissue temperature reached in the model. Since the tissue parameter functions used were approximated from measured values from the literature and these are only available below the boiling point of water, there may be deviations from reality here. This is especially the case when the boiling point of water is exceeded. An important next step is therefore to prove this by validating the simulations with experimental data. Further limitations are listed in Busch et al. [1]

IV. Conclusions

When considering the global current and the global tissue resistance, which result from solving the established FE model for coagulation of biological tissue by means of constantly applied DC voltage, the difference in the effects of change in thermal conductivity compared to the effects of change in electrical conductivity can be clearly shown. It must be further investigated to what extent these values can be used e.g., to realize a power control or for assessing the coagulation progress.

Author's Statement

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