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doi:10.15199/48.2020.01.12

Modelling the temperature-dependent RF ablation produced by the multi-tine electrode

***Abstract**. The paper discusses a thermal ablation of liver tissue with a multi-tine probe. The 3D model of such applicator with evenly distributed 2–5 arms was placed in hepatic tissue near a cylindrical vessel with circulating blood. The temperature of the surrounding tissue was regulated within defined limits by changing the voltage employed on the RF electrode. The current study examines influence of temperature-dependent and constant parameters of liver tissue on thermal efficiency of ablation procedure. As presented, a blood flowing in the blood vessel causes the cooling of tissue, which requires the electrode voltage to be increased to maintain the set therapeutic levels of temperatures. Interestingly, even larger changes in electrode potential are required in the case of temperature-dependent ablation with a blood vessel.*

***Streszczenie**. Praca omawia termiczną ablację tkanki wątroby za pomocą sondy wielopalcowej. Model 3D takiego aplikatora o równomiernie rozmieszczonych 2–5 ramionach umieszczono w tkance wątroby w pobliżu cylindrycznego naczynia z przepływającą krwią. Temperatura otaczającej tkanki była regulowana w określonych granicach poprzez zmianę napięcia przyłożonego na elektrodzie RF. W pracy zbadano wpływ zależnych od temperatury i stałych parametrów tkanki wątroby na sprawność ablacji cieplnej. Jak pokazano, krew płynąca w naczyniu krwionośnym powoduje chłodzenie tkanki, co wymaga zwiększenia napięcia elektrody, aby utrzymać zadane poziomy terapeutyczne temperatury. Co ciekawe, nawet większe zmiany potencjału elektrody są wymagane w przypadku zależnej od temperatury ablacji w obecności naczynia krwionośnego. (**Modelowanie zależnej od temperatury ablacji RF wytwarzanej przez elektrodę wielopalcową**)*

Keywords: temperature-dependent RF ablation, multi-tine electrode, hepatic tissue properties, blood flow, hepatocellular carcinoma, FEM.
Słowa kluczowe: ablacja RF, elektroda wielopalcowa, właściwości tkanki wątroby, przepływ krwi, rak komórkowo-wątrobowy, MES.

Introduction

Non-resectable liver lesions, such as hepatocellular carcinoma (HCC), are more and more often treated with intervention oncology, in particular using minimally invasive procedures, including radio frequency ablation (RFA), microwave ablation (MWA), laser induced interstitial (LITT) and others [1–4]. During thermal ablation treatment, various types of thin needle probes are placed in the hepatic tumor percutaneously and their position is monitored by means of ultrasound-, CT- or MR-guided imaging techniques. New applicators of various sizes and shapes, are designed [2], numerically tested [5, 6], and then verified during *in vitro* phantom [7] and *ex vivo* tissue [1] measurements to evaluate their performance and heating efficiency. What is important, the applicators employed in thermal therapies can be power- or temperature-controlled [2]. Thermal damage of tumor cells is based on denaturation of proteins, cell necrosis and coagulation of pathological tissues at high temperatures ranging from 50°C to 100°C [5].

Multifocal hepatic tumors due to complicated internal structure and dense vascularity should be modeled with regard to blood perfusion through the tissue as well as blood flow through the larger blood vessels in a liver [1, 2]. It should be noted that circulating blood in a vascular system causes the heat escaping from the live tissue, which requires higher applicator power to achieve the same thermal effect. Additionally, in order for the computer simulation for better reflection of actual tissue heating, the numerical analysis should take into account the real parameters of the liver tissue, and their frequency- and temperature-dependences as shown in some papers [8–12].

In this paper, based on some previously published papers [2, 5], a 3D model of multi-tine applicator for ablative heating was constructed. Next the optimal voltage values employed in RF electrode were established, that do not exceed set elevations of temperature valid in ablation treatment (50°C and 100°C). Importantly, a control parameter is temperature on the tip of principal tine-probe. The calculations have been done for the RF electrode with 2–5 evenly arrayed arms. Additionally, the effect of temperature-dependent parameters of hepatic tissue with and without circulating blood inside the vessel was examined. Related discussions can be found in many other papers concerning various types of RF electrodes for tumor ablation [13]–[17].

Model Definition and Governing Equations

The geometrical model of multi-tine electrode immersed in liver tissue is shown in Fig. 1. A special electric probe, designed to RF ablation of hepatic tumors, includes a cylindrical trocar made of stainless steel and several half-torus-shaped arms made of nickel-titanium. To conductive electrode, consisted of trocar tip and 2–5 tines, the electric voltage V_0 is applied. What is also important, the trocar is supplied from a 50-ohm feeding coaxial cable. All needed applicator dimensions were taken from the paper [2] and listed in Table 1. The RF electric probe was analyzed in a Cartesian coordinate system (x, y, z) centered inside a block of dimensions $10\ \mathrm{cm} \times 10\ \mathrm{cm} \times 12\ \mathrm{cm}$. What is more, near the RF electrode, at a distance of x_0 from the trocar axis, a cylindrical blood vessel with flowing blood is considered.

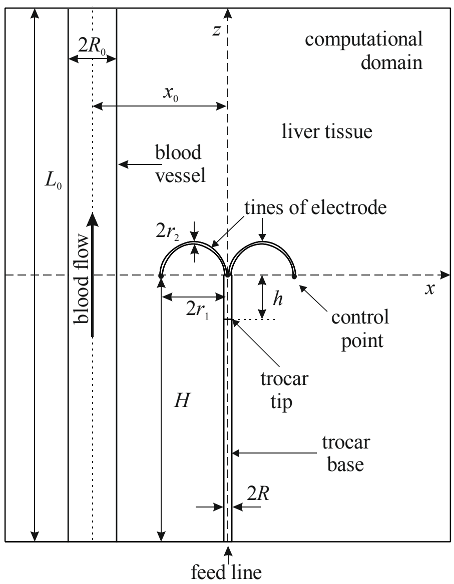

Fig.1. The *xz*-plane view of multi-tine probe for RF ablative heating

Table 1. Dimensions of RF applicator and blood vessel [5]

Elements	Size (mm)
radius of the trocar	$R = 0.9144$
outer radius of the electrode	$r_1 = 7.5000$
inner radius of the electrode	$r_2 = 0.2667$
length of the trocar	$H = 60$
length of the trocar tip	$h = 10$
radius of the blood vessel	$R_0 = 5$
length of the blood vessel	$L_0 = 120$
blood vessel center coordinate	$x_0 = 26$

For simplification, the model of tissue ablation assumes so-called quasi-static approximation for electro-conductive field according to the Laplace equation in the following form [5, 14]:

$$\nabla \cdot \left[-\sigma(T) \nabla \varphi \right] = 0 \quad (1)$$

where σ (S/m) corresponds to the temperature-dependent electric conductivity and φ (V) means an electric potential applied to the electrode.

A quasi-static approximation is fulfilled in the RF range $(450–550)$ kHz, since wavelength ($\lambda = c_0/f \approx 600$ m) is over 10,000 times larger than the largest dimension of the RF electric probe (Table 1) [14]. In such a case, the displacement currents are negligibly in comparison to the resistive currents and may be neglected [5]. In addition, all materials are modelled as an isotropic, homogeneous, and also linear media. Importantly, a liver tissue shows a dependence on temperature as described in the following section.

The bioheat transfer inside the hepatic tissue under RF ablation conditions can be evaluated from the Pennes equation as follows [5]:

$$\rho C(T) \frac{\partial T}{\partial t} + \nabla \cdot \left[-k(T) \nabla T \right] = Q_{\mathrm{b}} + Q_{\mathrm{ext}} + Q_{\mathrm{met}} \quad (2)$$

where ρ (kg/m^3), C (J/(kg·K)), and k (W/(m·K)) stand for the mass density, specific heat, and thermal conductivity, respectively. Moreover, T (K) is temperature and t (s)–time. Q_{b} (W/m^3) means volumetric power losses caused by blood perfusion of tissue, defined as [12]:

$$Q_{\mathrm{b}} = \rho_{\mathrm{b}} C_{\mathrm{b}} \omega_{\mathrm{b}}(T) \left[T_{\mathrm{b}} - T \right] \quad (3)$$

where T_{b} (K) denotes blood temperature, ω_{b} (1/s) – blood perfusion rate within liver tissue. Q_{ext} (W/m^3) indicates the so-called Joule heating [14, 18] specified by:

$$Q_{\mathrm{ext}} = \mathbf{J} \cdot \mathbf{E} = \sigma(T) \left| \mathbf{E} \right|^2 = \sigma(T) \left| \nabla \varphi \right|^2 \quad (4)$$

where $\mathbf{J}$ (A/m^2) and $\mathbf{E}$ (V/m) are vectors of current density and electric field strength. Moreover, Q_{met} (W/m^3) denotes a metabolic heat generation produced by the hepatic tissue.

Additionally, to estimate a blood flowing inside the blood vessel the Navier-Stokes (also called momentum balance) equation for incompressible and laminar liquid is employed [19]:

$$\rho_{\mathrm{b}} \frac{\partial \mathbf{u}}{\partial t} + \rho_{\mathrm{b}} \left(\mathbf{u} \cdot \nabla \right) \mathbf{u} - \eta \nabla^2 \mathbf{u} = -\nabla p \quad (5)$$

in combination with the mass balance law $\nabla \cdot \mathbf{u} = 0$, where $\mathbf{u}$ stands for the velocity vector of the venous blood, ρ_{b} (kg/m^3), η (Pa·s), p (Pa), denote mass density, dynamic viscosity and pressure of blood, respectively.

To solve described electro-thermal problem, the initial and boundary conditions for (1)–(5) should be specified. Generally, the electric insulation ($\partial\varphi/\partial n = 0$) was assumed on the external edge of the computational domain. What is more, the electric potential ($\varphi = V_0$) was applied on electrode surfaces including the trocar tip and tines of electrode (as depicted in Fig. 1). All utilized initial-boundary conditions are detailed described in some similar papers [5, 19].

Temperature-Dependent Tissue Parameters

Generally, an effective electrical conductivity of tissue consists of two terms, the static and alternating components, namely [10]:

$$\sigma_{\mathrm{eff}}(\omega) = \sigma_{\mathrm{s}} + \sigma_{\mathrm{AC}} \quad (6)$$

The static electrical conductivity σ_{s} is caused by the flow of free charges under constant electric field and influences the resistive (conductive) losses, which is assisted by the Joule heating. The alternating electrical conductivity σ_{AC} results from the rotation of electrical dipoles in AC electromagnetic field and affects the so-called dielectric losses of tissue. In most of biological structures, the resistive losses dominate in the case of stationary and alternating electric fields with frequencies vary up to several MHz. However, the dielectric losses are more dominant for microwave frequencies (above 300 MHz) and show the essential role in the tissue heating. In the range of microwave frequency the static conductivity can be successfully neglected [10].

Main tissue parameters, including electrical conductivity for the specified frequency of EM field, show temperature-dependent behaviour. Researchers in number amount of their numerical studies usually take into account the linear dependence of tissue conductivity in the form determined by performed measurements [8, 14]:

$$\sigma(T) = \sigma_0 \left[1 + \alpha_\sigma \left(T - T_{\mathrm{ref}} \right) \right] \quad (7)$$

where σ_0 stands for a baseline tissue conductivity measured at the reference temperature T_{ref} and α_σ means a temperature coefficient equal to the relative conductivity change per one degree of Kelvin or Celsius. What is important, the above relationship is valid for temperatures up to 100°C. At higher temperatures, the water in the tissue is transformed into gas and the tissue conductivity rapidly decreases to zero. In the case of *ex vivo* studies, the T_{ref}-parameter corresponds to the ambient room temperature at 20°C, and the α_σ-coefficient for hepatic tissue changes in the range $(0.006–0.02)$ K^{-1} [10].

A similar linear dependence on temperature can be also employed for estimating thermal conductivity of liver tissue at temperatures below 100°C [8, 13]:

$$k(T) = k_0 + \Delta k \left(T - T_{\mathrm{ref}} \right) \quad (8)$$

where k_0 indicates a baseline tissue thermal conductivity under the reference temperature T_{ref} and Δk represents the thermal conductivity increment for unit temperature, which for liver tissue varies from 0 to 0.0033 W/(m·K^2). In the case of *in vivo* and *in silico* studies, T_{ref} means the physiological temperature of body 37°C. Some researchers have shown that the adoption of a constant value of k-parameter may be also appropriate in numerical calculations because it has a limited influence on the results of RF ablation analysis [10].

At physiological temperatures, the specific heat capacity of liver tissue in a wide range of temperatures oscillates in the range of 3500–4000 J/(kg·K) [10]. Therefore, constant value of this parameter is very often assumed in numerical studies. However, the measurements presented in [9] show that the above limit temperature $T_{\mathrm{coag}} = 65$°C, at which a permanent coagulation of liver tissue occurs, a linear increase

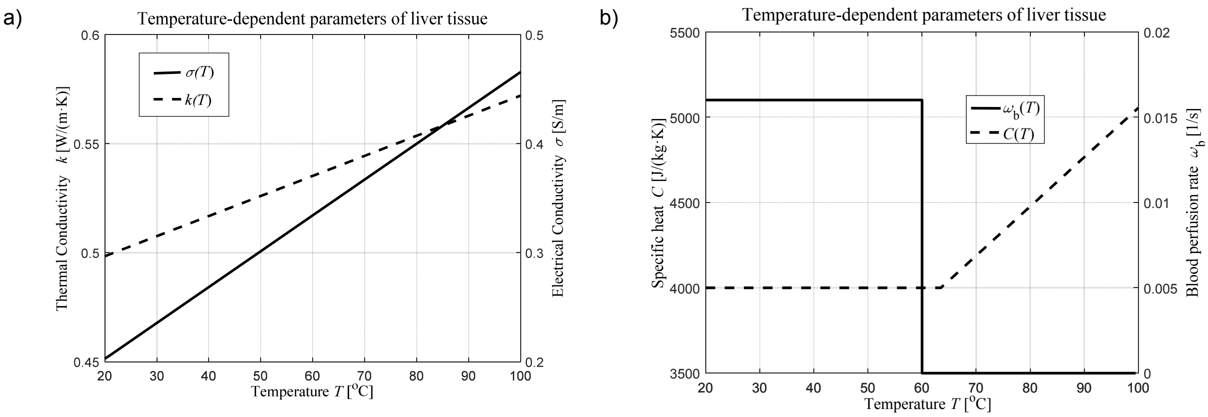

Fig.2. Distributions of temperature-dependent parameters: a) thermal and electrical conductivities, b) specific heat and perfusion rate of hepatic tissue

of this parameter is observed. For this reason, the following piecewise model of the C-parameter taking into account the specific heat of coagulated hepatic tissue was adopted [11]:

$$C(T) = \begin{cases} C_0, & T_b < T < T_{coag} \\ C_0 + \Delta C\left(T - T_{coag}\right), & T \geq T_{coag} \end{cases} \tag{9}$$

where C_0 denotes the baseline specific heat capacity of tissue and ΔC represents the specific heat increment under one degree $(1\ \mathrm{K})$ that for hepatic tissue is equal to $28.9\ \mathrm{J/(kg\cdot K^2)}$ [10]. Adopting the above model seems to take into account the real temperature-dependent behaviour of the liver tissue.

The blood flow through a tissue (the so-called perfusion) is an important element during RF ablation simulations, due to the dense vascularization of hepatic tissue. What is important, at high temperatures (above the limit temperature $T_{col} = 60°C$), as result of collapse of tissue vasculature, the blood clotting and tissue coagulation, the blood circulation and thereby the tissue cooling is stopped and the accumulation of heat inside a coagulated tissue occurs [5]. Taking the above into account, the blood perfusion rate for liver tissue as a temperature-dependent piecewise function is considered [4]:

$$\omega_b(T) = \begin{cases} \omega_{b0}, & T_b < T < T_{col} \\ 0, & T \geq T_{col} \end{cases} \tag{10}$$

All formulas of interest utilized in temperature-dependent model are summarized in Table 2 and illustrated in Fig. 2.

Table 2. The formulations for hepatic tissue parameters valid in simulation of RF ablation under 100°C [4, 10, 11, 13]

Quantity	Formula
electric conductivity	$\sigma(T) = 0.2028\left[1 + 0.0162\left(T - 20\right)\right]$
thermal conductivity	$k(T) = 0.503 + 0.00092\left(T - 25\right)$
specific heat capacity	$C(T) = \begin{cases} 3999.9, & 37°C < T < 63.5°C \\ 3999.9 + 28.9\left(T - 63.5\right), & T \geq 63.5°C \end{cases}$
blood perfusion	$\omega_b(T) = \begin{cases} 0.016, & 37°C < T < 60°C \\ 0, & T \geq 60°C \end{cases}$

Calculation Results

The described ablation model assumes so-called quasi-static approximation with the working frequency $f = 500\ \mathrm{kHz}$. The liver tissue, blood and electric probe parameters, employed for modelling non temperature-dependent RF ablation, have been taken from the similar papers [2, 5] and gathered in Table 3. What is more, the analyzed model assumes the blood perfusion rate equal to $\omega_b = 6.4\cdot10^{-3}\ 1/\mathrm{s}$, neglects the volumetric heat generation of liver tissue metabolism ($Q_{met} = 0$). In addition, the initial temperature is specified as $T_0 = T_b = 37°C$. All FEM-based calculations have been simulated in Comsol Multiphysics software. The analysis has shown that applied electrode voltage directly impacts the temperature distribution within hepatic tissue surrounding the electrode. First, the limit values of RF electrode voltage were established for two therapeutic levels of temperature valid during ablation treatment, namely $50°C$ and $100°C$. It was simply done by parametric changes of RF electrode voltage and monitoring the result temperature values in specified control point $(15.5,0,0)$ mm placed on the first tip of multi-tine electrode between the electrode and liver tissue as indicated in Fig. 1. As the author has shown in the previous work [5], due to the symmetrical construction of multi-armed applicator (where the angle ϕ in the xy-plane between individual tips is constant) and homogeneous excitation on all tips of the RF electrode (employed voltage is equal to V_0), temperature on the all tips of such electrode is almost the same ignoring the numerical errors. The temperature distributions depending on the electrode voltage for all analyzed models, including temperature-dependent behaviour of liver tissue parameters and presence of a blood vessel, are compared in Fig. 3a. Next, the transient temperature profiles of hepatic tissue were specified for the pre-established values of the multi-tine electrode voltages, namely $V_{0\,min}$, and $V_{0\,max}$, so the temperature levels in given control point does not exceed $50°C$ and $100°C$ (as shown in Fig. 3b). Interestingly, Table 4 gives established limit values of RF electrode voltage V_0 employed in different multi-tine probes in the cases of ablation with temperature-dependent and constant parameters and in the presence of blood flow and without it. Analysis also assumes that venous blood has following parameters: velocity of $u_z = 7.68\cdot10^{-4}$ m/s, and dynamic viscosity of $\eta = 3.5$ mPa·s, and the blood pressure difference at the ends of the vessel has zero value ($\Delta p = 0$).

Table 3. Parameters for modelling the RF ablation [2, 5]

Materials	σ (S/m)	ρ (kg/m^3)	C (J/(kg·K))	k (W/(m·K))
electrode	10^8	6 450	840	18
trocar tip	$4\cdot10^6$	21 500	132	71
trocar base	10^{-5}	70	45	0.026
liver tissue	0.333	1079	3540	0.520
blood	0.667	1000	4180	0.543

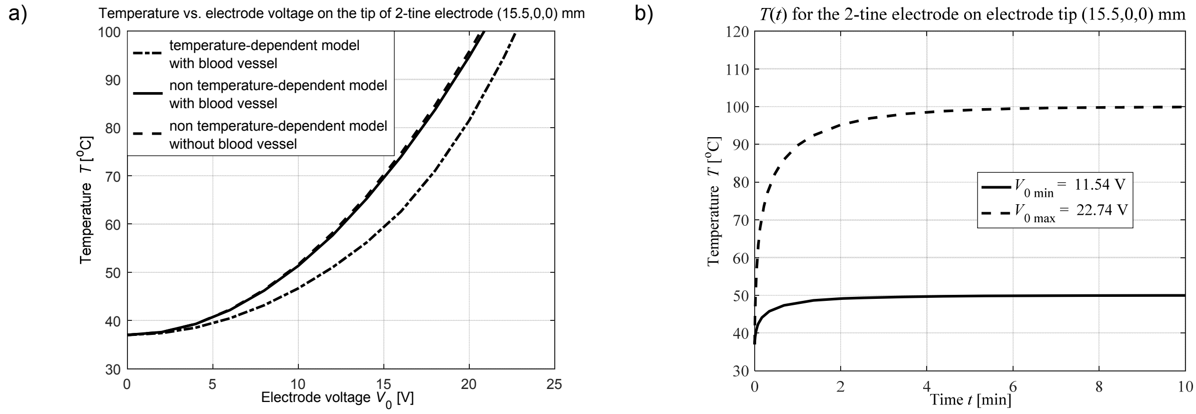

Fig. 3. Distributions of: a) temperature versus voltage on tip of the 2-tine electrode in the case all analyzed models, and b) temperature versus time on tip of the 2-tine electrode for two example voltage values employed on the 2-tine electrode immersed in the liver tissue during RF ablation procedure

Table 4. Limit voltage values employed on multi-tine electrodes

Limit values of electrode voltage	**2-tine electrode** ($\phi = 180°$)	**3-tine electrode** ($\phi = 120°$)	**4-tine electrode** ($\phi = 90°$)	**5-tine electrode** ($\phi = 72°$)
Non temperature-dependent model without blood vessel [5]				
$V_{0\,\min}$ (V)	9.40	9.78	10.19	10.58
$V_{0\,\max}$ (V)	20.68	21.52	22.41	23.28
Non temperature-dependent model with blood vessel				
$V_{0\,\min}$ (V)	9.48	9.93	10.34	10.71
$V_{0\,\max}$ (V)	20.87	21.79	22.69	23.59
Temperature-dependent model with blood vessel				
$V_{0\,\min}$ (V)	11.54	12.12	12.57	13.08
$V_{0\,\max}$ (V)	22.74	23.81	24.87	26.02

Summary

The current paper discusses the impact of temperature-dependent and constant parameters of hepatic tissue on the thermal efficiency of RF ablation procedure produced by the multi-tine electrode. As it was shown, relatively small differences in voltages of electrode not exceeding a value of $\Delta V_0 = 0.31$ V for $50°$C were observed between non temperature-dependent ablation models, containing the blood vessel and without it. As demonstrated, the blood flow in the blood vessel causes the tissue cooling, which requires the electrode voltage to be increased to maintain the temperature levels of interest. Even larger changes in electrode voltage up to $\Delta V_0 = 2.74$ V are required to achieve proper temperature profiles of tissue at $100°$C, in the case of a temperature-dependent ablation model including the blood vessel. This is associated with temperature growing of the basic parameters of coagulated liver tissue for temperature range up to $100°$C, as well as the blood clotting and retention of tissue perfusion for temperatures above $60°$C.

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