Design of a Novel Electrode of Radiofrequency Ablation for Large Tumors: A Finite Element Study

The aim of the study was to design a novel radiofrequency (RF) electrode for larger and rounder ablation volumes and its ability to achieve the complete ablation of liver tumors larger than 3 cm in diameter using finite element method. A new RF expandable electrode comprising three parts (i.e., insulated shaft, changing shaft, and hooks) was designed. Two modes of this new electrode, such as monopolar expandable electrode (MEE) and hybrid expandable electrode (HEE), and a commercial expandable electrode (CEE) were investigated using liver tissue with (scenario I) and without (scenario II) a liver tumor. A temperature-controlled radiofrequency ablation (RFA) protocol with a target temperature of 95 °C and an ablation time of 15 min was used in the study. Both the volume and shape of the ablation zone were examined for all RF electrodes in scenario I. Then, the RF electrode with the best performance in scenario I and CEE were used to ablate a large liver tumor with the diameter of 3.5 cm (scenario II) to evaluate the effectiveness of complete tumor ablation of the designed RF electrode. In scenario I, the ablation volumes of CEE, HEE, and MEE were 12.11 cm³, 33.29 cm³, and 48.75 cm³, respectively. The values of sphericity index (SI) of CEE, HEE, and MEE were 0.457, 0.957, and 0.976, respectively. The best performance was achieved by using MEE. In scenario II, the ablation volumes of MEE and CEE were 71.59 cm³ and 19.53 cm³, respectively. Also, a rounder ablation volume was achieved by using MEE compared to CEE (SI: 0.978 versus 0.596). The study concluded that: (1) compared with CEE, both MEE and HEE get larger and rounder ablation volumes due to the larger electrode–tissue interface and rounder shape of hook deployment; (2) HEE has the best performance in getting a larger and rounder ablation volume; and (3) computer simulation result shows that MEE is also able to ablate a large liver tumor (i.e., 3.5 cm in diameter) completely, which has at least 0.785 cm safety margin. [DOI: 10.1115/1.4038129]

Keywords: expandable electrode, finite element model, large ablation volume, radiofrequency ablation, sphericity index

Introduction

Radiofrequency ablation (RFA) as a thermal ablative method is used to treat focal primary and secondary malignancies in organs including the liver, lungs, kidneys, bones, and adrenal glands [1–3]. It has been shown that the clinical effectiveness of RFA drops off sharply for tumor diameters above 3 cm [4]. As a result, the clinical application of RFA is limited to small tumors (i.e., <3 cm in diameter) with a single electrode [5], called small tumor problem in the study. During the RFA procedure, tissue water boiling, vaporization, and carbonization usually start sequentially to occur when the tissue temperature is over 100 °C, which results in an immediate increase in impedance and subsequently a drop in the radiofrequency (RF) power that can be delivered [6,7].

To overcome the small tumor problem, we designed an expandable electrode to enlarge the interface between the tissue and electrodes to distribute the power to a wider zone [8]. So far, several similar expandable electrodes have been designed for the same reason, such as the umbrella electrode [8], the “Christmas tree” electrode [9], the InCircle Monarch electrode [10], the origami electrode [11], and the bicomponent conformal electrode [12]. Yet it is still a challenge to get a large (>3 cm in diameter) and round (close to a sphere) ablation zone for those expandable electrodes due to the shape and the difficulty in the deployment of hooks.

The purpose of the study was to evaluate the ablation results of our novel expandable electrode applied in two different modes compared to a commercial expandable electrode (CEE) in terms of the enlargement of the ablation volume in a spherical shape.

Materials and Methods

Design of a New RF Electrode. A 15-gauge CEE (RITA Medical Systems, Mountain View, CA) was chosen as a reference RF electrode [13]. The CEE contains four curved hooks that can be deployed from the electrode hollow in an umbrella-like shape (3 cm in diameter with full deployment) shown as in Fig. 1(a). In addition to the four hooks, the CEE also contains a 1-cm distal part as an active tip. Considering the shape of ablation zone generated by the CEE, an electrode with more spherical shape of deployed hooks and larger tissue-electrode interface was designed in the study. As shown in Fig. 1(b), the designed electrode has three parts, such as part A: six active hooks, part B: changing shaft, and part C: insulated part. The dimensions of the designed electrode are given in Fig. 1(b). For the designed electrode, there were two working modes considered in the present study: mode 1 (active tips: parts A and B, grounding pads: the side and bottom boundaries of liver tissue (Fig. 2). For this mode, the designed electrode was termed as monopolar expandable electrode (MEE))
and mode 2 (active tip: part A, grounding pads: part B) and the side and bottom boundaries of liver tissue. In this mode, the designed electrode was termed as hybrid expandable electrode (HEE)).

**Finite Element Model of RFA.** In the study, a three-dimensional finite element model (FEM) was used for RFA simulation. As shown in Fig. 2, a cylinder with 100 mm in diameter and 120 mm in height was expressed as the liver tissue. The RF electrode was placed at the center of liver with an insert depth of 60 mm.

The governing equation of the RFA procedure follows the Pennes’ bioheat transfer equation [1]

\[
\rho c \frac{\partial T(x, t)}{\partial t} = \nabla \cdot \left( k \nabla T(x, t) \right) + \rho_p c_p \omega_b (T_b - T(x, t)) + Q_m(x, t) + Q_b(x, t) \quad x \in \Lambda \tag{1}
\]

where \( \rho (\text{kg}^{-3}) \) is the density, \( c (\text{J kg}^{-1} \text{K}^{-1}) \) is the specific heat, \( T(x, t) (\text{C}) \) is the temperature, \( k (\text{W m}^{-1} \text{K}^{-1}) \) is the thermal conductivity, \( \rho_b (\text{kg}^{-3}) \) is the blood density, \( c_b (\text{J kg}^{-1} \text{K}^{-1}) \) is the specific heat of the blood, \( \omega_b (\text{s}^{-1}) \) is the blood perfusion, \( T_b \) is the temperature of the blood entering the tissue (assumed to be 37°C), \( x = \{x, y, z\} \) in the Cartesian coordinate system, \( \Lambda \) denotes the analyzed spatial domains, \( Q_m(x, t)(\text{W} \text{ m}^{-3}) \) is the energy generated due to metabolic processes which has been neglected since metabolic heat source is magnitude less than the spatial heat, and \( Q_b(x, t)(\text{W} \text{ m}^{-3}) \) generated by the RF electrical current as follows [14]:

\[
Q_b = J \cdot E = \sigma \nabla V^2 \tag{2}
\]

where \( J (\text{Am}^{-2}) \) is the current density, \( E (\text{V m}^{-1}) \) is the electrical field intensity, \( \sigma (\text{Sm}^{-1}) \) is the electrical conductivity, and \( V(V) \) is the applied voltage.

The values of the physical properties in the FEM adopted from the literature [9, 15, 16] are tabulated in Table 1. The present study assumed the thermal conductivity and electrical conductivity of both liver and tumor tissues at the temperature below 100°C as follows [17]:

\[
k(T) = k_0 + 0.0013(T(x, t) - T_0) \tag{3}
\]

\[
\sigma(T) = \sigma_0 [1 + 0.02(T(x, t) - T_0)] \tag{4}
\]

where \( k_0 \) and \( \sigma_0 \) are the thermal conductivity and electrical conductivity, respectively, measured at the temperature at 21°C (T0). Besides, when the temperature is above 100°C, the thermal conductivity was taken as a constant. The electrical conductivity decreases by 2 orders of the magnitude from 100°C to 105°C, and a constant value at the temperature above 105°C [15]. A piecewise function was used to describe the situation of blood perfusion in liver and tumor tissues [18]

\[
\omega_b(t) = \begin{cases} 
\omega_{b0} & \text{for } T \leq 50^\circ \text{C} \\
0 & \text{for } T > 50^\circ \text{C} 
\end{cases} \tag{5}
\]

where \( \omega_{b0} \) is the constant blood perfusion of tissue (refer to Table 1).

In the present study, the induced thermal damage was evaluated by the Arrhenius model for the estimation of ablation zones [19]

\[
\Omega(t) = A \int_0^\infty e^{\frac{-m\Delta E}{kT}} dt \tag{6}
\]

where \( A (\text{s}^{-1}) \) is the frequency factor, \( \Delta E (\text{J mol}^{-1}) \) is the activation energy barrier, \( R (\text{J mol}^{-1} \text{K}^{-1}) \) is the universal gas constant (8.314 J mol^{-1} K^{-1}), and \( T(\text{K}) \) is the absolute temperature. For the liver and tumor tissues, \( A = 7.390 \times 10^{10} \text{ s}^{-1} \) and \( 3.247 \times 10^{13} \text{ s}^{-1} \), respectively, and \( \Delta E = 2.577 \times 10^3 \text{ J mol}^{-1} \) and \( 2.814 \times 10^3 \text{ J mol}^{-1} \), respectively [20]. The threshold of tissue death is \( \Omega(t) = 1 \), which means a 63% probability of cell death (D63) [21].

**Table 1 Physical properties of the materials used in the model**

<table>
<thead>
<tr>
<th>Property</th>
<th>Liver</th>
<th>Tumor</th>
<th>Active tip</th>
<th>Insulated shaft</th>
<th>Blood</th>
</tr>
</thead>
<tbody>
<tr>
<td>( \rho (\text{kg}^{-3}) )</td>
<td>1080</td>
<td>1045</td>
<td>6450</td>
<td>70</td>
<td>1000</td>
</tr>
<tr>
<td>( c (\text{J kg}^{-1} \text{K}^{-1}) )</td>
<td>3455</td>
<td>3760</td>
<td>840</td>
<td>1045</td>
<td>4180</td>
</tr>
<tr>
<td>( k (\text{W m}^{-1} \text{K}^{-1}) )</td>
<td>0.515 (a)</td>
<td>0.60 (a)</td>
<td>18</td>
<td>0.026</td>
<td>0.49</td>
</tr>
<tr>
<td>( \sigma (\text{S m}^{-1}) )</td>
<td>0.203 (a)</td>
<td>0.50 (a)</td>
<td>1.0 (\times) 10(^8)</td>
<td>1.0 (\times) 10(^{-5})</td>
<td>0.667</td>
</tr>
<tr>
<td>( \omega_b (\text{s}^{-1}) )</td>
<td>0.016 (a)</td>
<td>0.002 (a)</td>
<td>——</td>
<td>——</td>
<td>——</td>
</tr>
</tbody>
</table>

\(a\)Evaluated at 21°C.

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Fig. 1 Dimensions (in mm) and structures of two RF electrodes: (a) CEE and (b) the designed electrode.

Fig. 2 Three-dimensional geometric model (in mm) of liver tissue with the liver tumor and the designed RF electrode.

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Transactions of the ASME
Table 2: Number of elements for each scenario

<table>
<thead>
<tr>
<th>Scenario</th>
<th>Model</th>
<th>Number of elements</th>
</tr>
</thead>
<tbody>
<tr>
<td>One-compartment model</td>
<td>CEE</td>
<td>231,483</td>
</tr>
<tr>
<td></td>
<td>MEE</td>
<td>426,934</td>
</tr>
<tr>
<td></td>
<td>HEE</td>
<td>426,934</td>
</tr>
<tr>
<td>Two-compartment model</td>
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</tr>
<tr>
<td></td>
<td>MEE</td>
<td>426,934</td>
</tr>
</tbody>
</table>

**Initial and Boundary Conditions.** The initial temperature of the whole tissue domain was set at 37°C (inner body temperature). A zero-voltage electric potential was applied at the grounding pads. The boundaries of the active tip were set to a variable power source determined by a proportional–integrative–derivative controller (Kp = 32, Ki = 1, and Kd = 0.001). The upper boundary of liver tissue was set as electrically isolative. The other electrical and thermal boundary conditions of the FEM were considered as continuous.

**Mesh Convergence Test.** The software of COMSOL MULTIPHYSICS 5.2a (COMSOL, Inc., Burlington, MA) was used to simulate and predict the ablation zone. The CEE was used to perform the mesh convergence test. In the convergence test, RFA was applied at the constant applied voltage of 30 V and the ablation time of 200 s in scenario I. The mesh convergence criterion was considered as the difference in the maximum temperature between two contiguous meshes smaller than 0.1%. The free tetrahedral mesh was used to discretize the physical domains of the model, which was divided as two parts: part I (active tips, insulated shaft, and the tumor sphere) and part II (the remaining parts of the model). The mesh of part II was fixed at the default “normal” in COMSOL; the mesh of part I was set as the default “finer” with the minimum element size of 0.5 mm and further refined until convergence criterion got satisfied. Then, the previous mesh with fewer elements was used to reduce the computational load. The meshing results of elements are given in Table 2.

**Validation of the FEM Model.** The results of in vivo or ex vivo experiments of RFA using the 15-gauge CEE from Ref. [13] were chosen as the references. The same situations with the work of de Baere et al. [13] were considered in the validation, such as the dimensions of RF electrode, boundary conditions, and anatomical environments. However, it is worth mentioning that the human physical properties of liver tissue were used in the study, unlike the porcine and bovine studies of de Baere et al. Both in the work of Hall et al. [22] and Trujillo and Berjano [15] concluded that properties affecting the ablation volume in computer simulation significantly are the electrical conductivity and perfusion ratio of liver tissue. By examining the liver electrical conductivities of human, pigs, and cows (0.467–0.512, 0.465–0.528, and 0.488 S m⁻¹, respectively [22–26]) and the perfusion ratios of the human and porcine liver (0.009–0.019 and 0.018 s⁻¹, respectively [22,27]), we believe that the work of de Baere et al. is reliable to be used as the reference.

The ablation volume was calculated approximately by using the following equation:

\[ V = \pi \left( d_1 \cdot d_2 \cdot d_3 \right) / 6 \]  

(7)

where \( d_1 \) and \( d_2 \) are the diameters perpendicular to the axis of the electrode, and \( d_3 \) is the diameter along it. In the literature, de Baere et al. assumed that \( d_1 = d_2 \) due to the technical difficulty of measurement. For the ablation size of the FEM, the ablation volume was calculated by using Eq. (6) \((\Omega(t) \geq 1)\), while \( d_1 \), \( d_2 \), and \( d_3 \) were measured by the software along the directions mentioned previously. As shown in Table 3, there was good agreement between ablation size found for FEM and the experiments. It is also worth mentioning that \( d_1 \) (in vivo and ex vivo) in the FEM is slightly larger than its peer in the experiment. The reasons for this might be the measurement error in the experiment and the blood perfusion error in simulation. Nonetheless, it is reasonable to accept the proposed FEM of RFA as a test-bed for ablation volume estimation.

**Analysis of Sphericity.** In the present study, the sphericity of the ablation volume was also investigated. A near-spherical shape is desirable because most tumors are more or less round [28]. Clinically, it is important to know the shape of the proposed ablation zone so that the electrodes can be placed in such a way as to optimize the ablation zone relative to the tumor. A sphericity index (SI) was used to assess the shape of the ablation volumes for the three RF electrodes [29]

\[ SI = \frac{V}{\pi \left( \max(d_1, d_2, d_3) \right)^3 / 6} \]  

(8)

where \( \max(d_1, d_2, d_3) \) is the maximum value among \( d_1 \), \( d_2 \), and \( d_3 \).

**Study Design.** A proportional–integrative–derivative controller was implemented in the COMSOL software to control the temperature at a hook tip (Fig. 1). The goal was to achieve a temperature of 95°C for 15 min [13]. As mentioned before, two scenarios were considered in the present study. Scenario I: three RF electrodes (i.e., CEE, MEE, and HEE) were compared in terms of size and the sphericity of the ablation volume using one-compartment RFA model. Scenario II: the electrode that achieved the largest ablation volume and the highest SI was compared to the CEE in the treatment of a liver tumor with a diameter of 3.5 cm. Successful ablation was defined using the clinical standard of complete ablation of the liver tumor and ablation of a 0.5–1 cm margin of normal tissue surrounding the tumor.

**Results**

**Results of Scenario I.** Figures 3(a)–3(c) show the potential distribution of CEE, HEE, and MEE at the end of treatment (15 min), respectively. The temperature distribution of CEE, HEE, and MEE is shown in Figs. 3(d)–3(f), respectively. It is worth mentioning that the high-temperature areas (>90°C) are different among the three electrodes, as shown in Figs. 3(d)–3(f). For CEE and MEE, the maximum temperature (i.e., 95.5°C and 99.6°C, respectively) was at the area around the hooks. For HEE, however, the maximum temperature (i.e., 106.7°C) was at the area around part B of the electrode. Figure 4 shows the results of ablation zone generated by the three RF electrodes (i.e., CEE, HEE, and MEE). From Fig. 4, we can find that the ablation zones generated by HEE and MEE were much better than that generated by CEE in both the ablation volume and shape. The ablation volumes of CEE, HEE, and MEE were 12.11 cm³, 33.29 cm³, and 48.75 cm³, respectively. Compared with CEE, the ablation volumes of HEE and MEE increased by 174.9% and 302.6%, respectively. The same conclusion was also able to be found in

Table 3: Comparisons of ablation size from the FEM and ex vivo/in vivo experiments

<table>
<thead>
<tr>
<th>FEM</th>
<th>Ex vivo</th>
<th>In vivo</th>
<th>Experiment</th>
</tr>
</thead>
<tbody>
<tr>
<td>( d_1 ) (cm)</td>
<td>4.51</td>
<td>3.70</td>
<td>3.94 ± 0.38</td>
</tr>
<tr>
<td>( d_2 ) (cm)</td>
<td>3.94</td>
<td>2.55</td>
<td>3.94 ± 0.38</td>
</tr>
<tr>
<td>( d_3 ) (cm)</td>
<td>3.82</td>
<td>2.75</td>
<td>3.46 ± 0.5</td>
</tr>
<tr>
<td>( V ) (cm³)</td>
<td>34.20</td>
<td>12.11</td>
<td>26.67 ± 9.59</td>
</tr>
</tbody>
</table>
the shape of ablation volume. As shown in Table 4, the ablation volumes generated by HEE and MEE were very close to spheres (SI = 0.957 and 0.976, respectively). However, the ablation volume generated by CEE was in an umbrella shape with SI = 0.457. MEE was found to be the best performing electrode with the ablation volume of 48.75 cm$^3$ and SI = 0.976. Thus, MEE was chosen to be used in scenario II.

**Results of Scenario II.** In scenario II, a liver tumor with the diameter of 3.5 cm was taken into account. CEE and MEE were used to treat the liver tumor in a clinical way (15 min and temperature-controlled RFA at 95°C). Table 5 shows the results of ablation zones and sphericity indices generated by CEE and MEE. Compared with CEE, the sizes of ablation zone generated by MEE were increased by 59.4%, 30.7%, and 61.8% in the diameters of $d_1$, $d_2$, and $d_3$, respectively. The ablation volume was increased by 266.6% from 19.53 cm$^3$ to 71.59 cm$^3$. More spherical ablation zone was achieved by using MEE (SI = 0.978) compared with CEE (SI = 0.596). Furthermore, at least 0.785 cm ablative margin surrounding the liver tumor can be obtained in all the diameters of $d_1$, $d_2$, and $d_3$, as shown in Table 5 and Fig. 5.
Table 4  Ablation volumes and SIs of three electrodes

<table>
<thead>
<tr>
<th>Electrode</th>
<th>V (cm³)</th>
<th>SI</th>
</tr>
</thead>
<tbody>
<tr>
<td>CEE</td>
<td>12.11</td>
<td>0.457</td>
</tr>
<tr>
<td>HEE</td>
<td>33.29</td>
<td>0.957</td>
</tr>
<tr>
<td>MEE</td>
<td>48.75</td>
<td>0.976</td>
</tr>
</tbody>
</table>

Table 5  Ablation results of CEE and MEE for a liver tumor with 3.5 cm in diameter

<table>
<thead>
<tr>
<th>Electrode</th>
<th>d¹ (cm)</th>
<th>d² (cm)</th>
<th>d₃ (cm)</th>
<th>V (cm³)</th>
<th>SI</th>
<th>Complete ablation</th>
</tr>
</thead>
<tbody>
<tr>
<td>CEE</td>
<td>3.97</td>
<td>3.18</td>
<td>3.14</td>
<td>19.53</td>
<td>0.596</td>
<td>No</td>
</tr>
<tr>
<td>MEE</td>
<td>5.19</td>
<td>5.07</td>
<td>5.08</td>
<td>71.59</td>
<td>0.978</td>
<td>Yes</td>
</tr>
</tbody>
</table>

Fig. 5  Tumor and liver tissue death rates in scenario II: section view crossing the axis of (a) CEE and (b) MEE, respectively, and section view perpendicular to the axis of (c) CEE and (d) MEE, respectively, at the dashed line (the black circle means the liver tumor)

Discussion

In scenario I, both HEE and MEE can achieve larger and rounder ablation volumes compared to CEE. We attribute this to the larger electrode–tissue interfaces and the rounder shape of hooks. Furthermore, a larger and rounder ablation volume can be achieved by MEE compared to HEE (ablation volume: 48.75 cm³ versus 33.29 cm³ and SI: 0.976 versus 0.957). This phenomenon may be attributed to the following two reasons. The first reason is the role of part B of the designed electrode. In the case of HEE, the alternating current not only flows to the tissue bottom and side boundaries but also flows to part B of the designed electrode, as shown in Fig. 1(b). A smaller potential difference at the ablation boundaries can be found in HEE compared to MEE, shown in Figs. 3(b) and 5(c). Such a difference is due to a reduction in power delivered to the tissue areas outside the hooks, which results in a smaller ablation diameter perpendicular to the electrode compared with MEE. However, there is no significant difference in the ablation diameter along the electrode. Therefore, a larger and rounder ablation volume is achieved in MEE. The second reason is the position of the maximum temperature. For HEE, the maximum temperature is found in the area surrounding part B, which is different from MEE where the maximum temperature is around the hooks. This phenomenon is because of the tremendously Joule heat generated by RF power around part B due to the larger potential difference. Therefore, a greater amount of RF power is delivered to the tissue areas inner the hooks in HEE than MEE.

It is noticed that the maximum temperature (106.7 °C) occurred in the area surrounding part B in the case of HEE, while the temperature at one of the hook tips was controlled at 95 °C. This phenomenon was believed as the reason of dramatically decrease in potential in the area surrounding part B that was set as grounding pad (Fig. 3(b)). Thus, the tissue dehydration might be the case in the same area, which can lead to a certain degree of error in the calculation of ablation volume. Due to the target temperature (95 °C) and the ablation time used in the study, only a very limited area of tissue surrounding part B was dehydrated (Fig. 3(e)). So the error in ablation volume should be negligible in the study. Otherwise, it is necessary to consider the tissue phase change in the FEM as it affects the accuracy of ablation volume.

In scenario II, due to the tumor characteristics, both HEE and MEE generate larger ablation volumes than that in scenario I. For CEE, only a fan-shaped ablation zone is generated along the section view of cross-electrode-axis (Fig. 5(a)), which cannot completely cover the tumor area. Because there are only four hooks on this commercial electrode, the size of ablation zone between the two adjacent hooks (d₂) is much smaller than the size between the two diagonal hooks (d₁) (Fig. 5(c)). And the semicircle hooks of CEE cannot achieve a sphere ablation volume (Fig. 5(a)). As shown in Figs. 5(b) and 5(d), MEE generates the ablation volume with at least 0.785 cm safety margin, which supports the assertion that our novel monopolar electrode has the potential to ablate tumors larger than 3 cm completely.

A limitation of this preliminary study is the lack of experimental results. Future studies are warranted to the fabrication of the design RF electrode and its application using tissue-mimicking phantom followed by ex vivo/in vivo porcine liver tissue. The difficulty in the deployment of hook can be imagined due to its spherical and long shape. Improvements on the design of hook or the analysis of forces between hook and biological tissue during its deployment must be investigated in the future work.

Conclusion

A novel expandable electrode was designed and assessed using finite element model in the study, and the following three conclusions can be drawn:

1. Compared with CEE, both MEE and HEE can achieve larger and rounder ablation volumes due to the larger electrode–tissue interface and the rounder shape of hooks.
2. Compared with HEE, MEE yielded a larger and rounder ablation volume.
3. MEE was able to completely ablate a large liver tumor (i.e., 3.5 cm in diameter) with at least 0.785 cm safety margin.

Funding Data

- National Sciences and Engineering Research Council of Canada (NSERC) through a discovery grant to Wenjun Zhang.
- National Natural Science of China (Grant No. 51375166).

Nomenclature

- A = frequency factor
- c = specific heat of blood
- cₙ = specific heat of liver
- CEE = commercial expandable electrode
- dₙ = longest diameters perpendicular to the axis of the electrode
\[ d_2 = \text{shortest diameters perpendicular to the axis of the electrode} \]

\[ d_1 = \text{diameter along the electrode} \]

\[ E = \text{electrical field intensity} \]

\[ \text{FEM} = \text{finite element model} \]

\[ \text{HEE} = \text{hybrid expandable electrode} \]

\[ f = \text{current density} \]

\[ k = \text{thermal conductivity of liver tissue} \]

\[ k_0 = \text{coefficient for the derivative term} \]

\[ k_1 = \text{coefficient for the integrative term} \]

\[ k_2 = \text{coefficient for the proportion term} \]

\[ \Omega = \text{target tissue size in liver tumors using two-compartment finite element model} \]

\[ Q_b = \text{energy generated by the RF electrical current} \]

\[ Q_m = \text{energy generated due to metabolic processes} \]

\[ \mathcal{R} = \text{universal gas constant} \]

\[ \text{RF} = \text{radiofrequency} \]

\[ \text{RFA} = \text{radiofrequency ablation} \]

\[ \text{SI} = \text{sphere index} \]

\[ t = \text{ablation time} \]

\[ T = \text{temperature of liver} \]

\[ T_0 = \text{temperature of the blood entering the tissue} \]

\[ U = \text{applied voltage} \]

\[ V = \text{ablation volume} \]

\[ x = \text{Cartesian coordinate system} \]

\[ \Delta E = \text{activation energy barrier} \]

\[ \rho = \text{density} \]

\[ \rho_b = \text{blood density} \]

\[ \sigma = \text{electrical conductivity of liver tissue} \]

\[ \sigma_0 = \text{electrical conductivity measured at 21°C} \]

\[ \Omega = \text{degree of tissue death} \]

\[ \omega_b = \text{blood perfusion} \]

\[ \omega_m = \text{constant blood perfusion of tissue} \]

\[ \wedge = \text{analyzed spatial domains} \]

**References**


