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Experimental and Theoretical Study of an Internally Cooled Bipolar Electrode for RF Coagulation of Biological Tissues

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Abstract—Although some types of bipolar electrodes have been broadly employed in clinical practice to coagulate biological tissue by means of radiofrequency (RF) currents, there is still scanty available information about their electrical-thermal behaviour. We are focused on internally cooled bipolar electrodes. The goal of our study was to know more about the behavior of this kind of electrodes. For that, we planned an experimental and theoretical model. The experimental study was based on bovine hepatic ex vivo tissue and the theoretical model was based on the Finite Element Method (FEM). In order to check the feasibility of the theoretical model, we assessed both theoretically and experimentally the effect of the internal cooling characteristics of the bipolar electrode (flow rate and coolant temperature) on the impedance progress during RF heating and coagulation zone dimensions. The experimental and theoretical results were in good agreement, which suggests that the theoretical model could be useful to improve the design of cooled bipolar electrodes.

I. Introduction

LIVER resection remains a major procedure carrying a significant risk of intraoperative bleeding leading to blood transfusion which has been shown to affect postoperative morbidity, mortality and long-term survival. Therefore, over the past decade many techniques have been developed to minimize intraoperative haemorrhage during this type of operation, such as the ultrasonic dissector, the WaterJet dissector, or stapling devices. Some methods employed radiofrequency (RF) heating to achieve coagulate the liver tissue before resecting it [1]. This kind of energy is applied to the tissue by means of electrodes. In particularly, bipolar electrodes are composed of two identical electrodes between which RF current is flowing. The use of bipolar electrodes prevents RF currents flow through other tissues, thus minimizes the risk of injury to adjacent tissues. This is especially important when surgery is performed by laparoscopic approach where the visibility is reduced [2]. Moreover, it is known that the thermal cooling inside the RF electrode (cooled electrode) avoids dehydration and carbonization (temperatures ≈100ºC) of tissue adjacent to the electrode. This allows RF currents continue flowing to deeper areas in the tissue, and hence the lesion (coagulation) depth is increased, minimizing the bleeding.

Up to date, bipolar electrodes have been broadly employed to coagulate blood vessels. More recently, other more sophisticated bipolar electrodes have been proposed to conduct more specific surgical procedures. For instance, the cooled bipolar electrode of Atricure (West Chester, OH, USA) is employed to coagulate cardiac tissue in order to treat atrial fibrillation [3]. Although this type of electrode has been used in clinical practice, little information about their electrical-thermal behavior is available. For this reason, the aim of our study was to know more about that behavior. In particular, we planned an experimental and theoretical model. The experimental study was based on bovine hepatic ex vivo tissue. On the other hand, the theoretical model was based on the Finite Element Method (FEM). Moreover, in order to check the feasibility of the theoretical model, we assessed the effect of the internal cooling characteristics of the bipolar electrode on the impedance progress during RF heating and coagulation zone dimensions.

II. Materials and Methods

A. Experimental study

The device (Fig. 1) consists of two identical electrodes (5 mm diameter, 0.25 wall thickness) separated a distance of 6 mm. The internal cooling is conducted by means of a peristaltic pump (Watson Marlow, Wilmington, MA, USA).

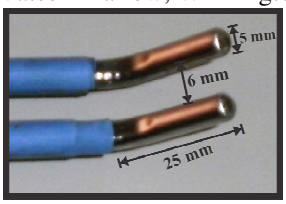

Fig. 1. Cooled bipolar electrode employed in our study.

The electrodes are connected to a RF generator CC-1 Cosman Coagulation System (Radionics, Burlington, MA, USA), which delivers a non modulated sinusoid waveform up to 100 V (rms) on a 100 Ω and a maximum current of 1 A. The electrical variables voltage and impedance were recorded by the RF generator and processed them by means of Agilent VEE software (Agilent Technologies, Santa Clara, CA) with a sampling rate of 30 Hz.

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Thirty-six coagulations were performed over the surface of a beef liver (4.5 kg weight, 17ºC initial temperature). We assessed the effect of changing the flow rate of the circulating fluid and the coolant temperature. We considered three flow rates (50, 75 and 100 mL/min) and two coolant temperatures (20ºC and 5ºC). A total of six lesions were conducted in each case by setting a constant voltage of 50 V (rms) for 60 s at the RF generator. The room temperature was 22ºC. We analyzed the impedance progress during RF heating and the coagulations created in the tissue after heating. A slice of each lesion was transversally cut in order to characterize its geometry (see Fig. 2 (a)). The coagulation zone geometry was quantified by the white coagulation contour and, it was characterized by its depth (D) and width (W) in the tissue (see Fig. 2 (b)). Both parameters were displayed as the mean ± standard deviation. A Mann-Whitney Test was used to analyze the differences between these experimental cases. Data collection and analysis of the impedance progress were performed with Matlab® (The MathWorks, Natick, MA, USA) and statistical software (SPSS 17.0, Chicago, IL, USA). Moreover, in order to relate this lesion contour with a particular isothermal line in the theoretical model, we assessed the change in the color of the liver tissue by immersing it in heated water during a certain time. We previously changed the water temperature from 50 to 75ºC and the time from 15 to 60 s. When the water temperature was 70ºC, the tissue assumed a white color, even for 15 s. Therefore, we considered the 70ºC isotherm as thermal lesion marker in all computer simulations.

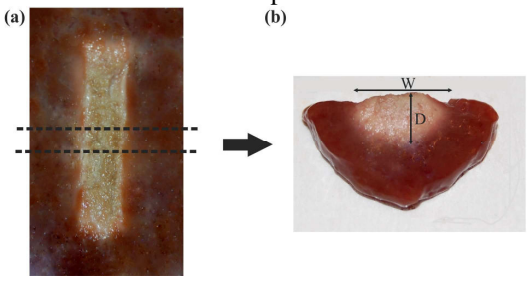

Fig. 2. Coagulations created in the tissue: (a) surface and (b) side view, in which is represented the lesion parameters: depth (D) and width (W). The white coagulation determined the lesion contour.

B. *Theoretical model*

The theoretical problem was based on a coupled electric-thermal problem, which was solved numerically using the FEM by means of COMSOL Multiphysics software (COMSOL Inc., Burlington MA, USA). Fig. 3 shows the proposed theoretical model, which represents the device over a fragment of hepatic tissue. Since there is a symmetrical plane, the model only includes the half of all electrodes-tissue. The electrode with 5 mm diameter is assumed to be inserted in the tissue a depth of 0.5 mm and is separated from symmetrical plane a distance of 3 mm (6 mm inter-electrode distance). The tissue dimensions R and H were estimated by means of a sensibility analysis in order to avoid boundary effects. A convergence test was performed to obtain the adequate spatial (i.e. minimum meshing size) and temporal resolution, as the same way as in [4]. The value of the maximal temperature achieved in the tissue (T_{max}) after 60 s of heating was used as a control parameter. As a result sensitivity analyses, we obtained the following tissue dimensions: R = H = 50 mm. The convergence test provided a grid size of 0.115 mm in the electrode-tissue interface, and a step time of 0.05 s.

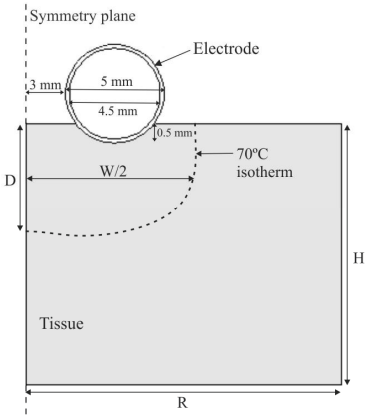

Fig. 3. Theoretical model proposed (out of scale). Depth (D) and width (W) of the lesion were assessed using the 70ºC isotherm.

The thermal and electrical characteristics of the model elements were obtained from [5]-[6]. The electrical and thermal conductivity of the tissue were piecewise functions. For the electrical conductivity (σ) we considered an exponential growth of 1.3%/ºC up to 100ºC. This value was obtained by varying it until achieve the same slope than experimentally. Then it was kept constant between 100 and 105ºC and finally σ decreased linearly 2 orders for five degrees. The thermal conductivity (k) grew linearly $0.0015ºC^{-1}$ up to 100ºC, after which temperature k was kept constant [7]. Analytically, the piecewise functions were defined using the Heaviside function, which in COMSOL was substituted by the smoothed function *flc2hs* [8].

Simulations were conducted using a constant electrical voltage of 25 V on the electrode and 0 V at the symmetry plane. This value models an RF heating of 50 V (rms). A null electrical current was used on the surfaces at a distance from the electrode, and on the tissue-ambient and electrode-ambient. Regarding the thermal boundary conditions, null thermal flux was used at the symmetry plane. The temperature for surfaces at a distance from the electrode was 17ºC (equal to the experimental conditions). The effect of free convection at the tissue-ambient and electrode-ambient interfaces was taken into account using a thermal transfer coefficient (h_e) of 20 W/m^2K, and the room temperature was considered to be 22ºC. The thermal boundary condition of convective coefficient (h_i) was applied at the inner electrode part in order to model the cooling effect of the circulating fluid. The value of h_i for laminar flow, was calculated as described in [4]. Thus, we obtained three values of h_i: h_i of 1134, 1336 and 1537 W/m^2K for a flow rate of 50, 75 and 100 mL/min, respectively. The coolant temperature was 20ºC and 5ºC.

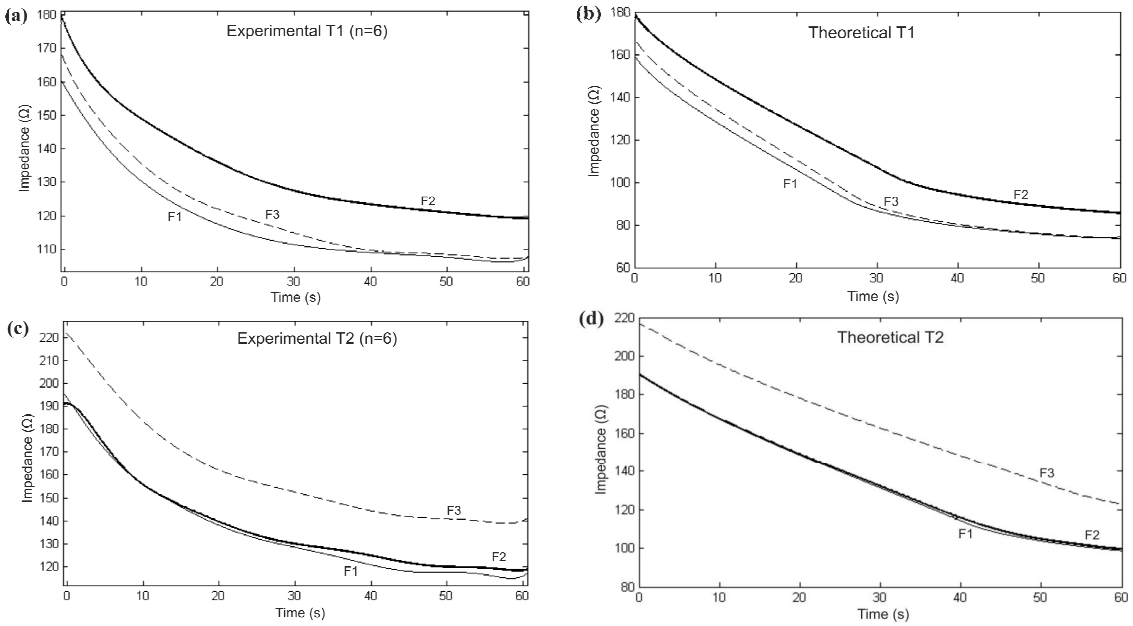

Fig. 4. Impedance progress obtained from the experimental ((a) and (c)) and theoretical model ((b) and (d)) by changing flow rate (F1: 50 mL/min, F2: 75 mL/min, F3: 100 mL/min) and coolant temperature (T1: 20ºC, T2: 5ºC).

As we have mentioned above, the model was based on a coupled electric-thermal problem. The governing equation for the thermal problem is the Bioheat Equation [7]:

$$\rho \cdot c \cdot \frac{\partial T}{\partial t} = \nabla(k\nabla T) + q + Q_p + Q_m \qquad (1)$$

where T is temperature, t is time, ρ is density, c is specific heat and k is thermal conductivity. The term Q_p corresponds with heat loss from blood perfusion and Q_m is metabolic heat generation. This last term is ignored in the RF of the liver since it has been shown to be insignificant. Since the experimental study was ex vivo, Q_p was not also considered. Finally, the term q is heat source caused by RF power (Joule loss) which is given by $q = \sigma \cdot |E|^2$, where $|E|$ is the magnitude of the vector electric field (V/m). The value of this vector is evaluated from $\vec{E} = -\nabla\Phi$, where Φ is the voltage (V). The voltage is obtained by using Laplace's equation $\nabla \cdot \sigma\nabla\Phi = 0$, which is the governing equation of the electrical problem. We used a quasi-static approach due to the frequencies used in RF (≈ 500 kHz) and for the geometric area of interest the tissues can be considered as purely resistive [7]. We considered hepatic tissue vaporization and we modeled this phenomenon using the enthalpy method according to [9].

III. Results

A. Impedance progress

Fig. 4 shows the experimental and theoretical impedance progress obtained by changing flow rate and coolant temperature. For the experimental part, we considered the mean of six coagulations per case. In order to compare theoretical and experimental results, we adjusted the initial impedance of the tissue in the theoretical model. This was done by varying the initial tissue electrical conductivity to achieve the same initial impedance than experimentally. Such as observed in Fig. 4, the impedance progress from theoretical model and experiments followed a similar trend for any of the studied cases. Indeed, the impedance decreased until a specific time, after which it remained practically constant until the end-point time. That specific time was reached earlier in the case of lower flow rates and higher cooling temperatures. For instance, for 20ºC coolant temperature and 50 mL/min flow rate (see Fig. 4 (a) y (b)), the impedance decreased from an initial value of ≈ 160 Ω until ≈ 110 Ω in the experiments and to ≈ 95 Ω in the theoretical model, both at ≈ 30 s. After that, impedance remained constant until 60 s.

B. Coagulation zone dimensions

Fig. 5 shows the experimental and theoretical coagulation zone geometry created in the tissue after RF heating, considering a 100 mL/min flow rate and a 20ºC coolant temperature. Note the similarity between experimental and theoretical results, which was repeated in all the cases.

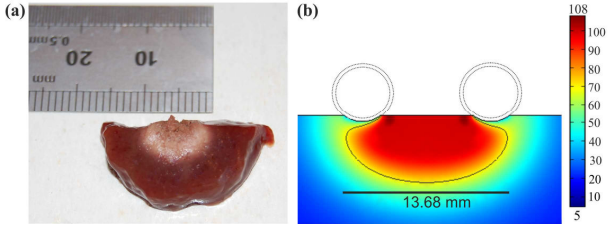

Fig. 5. Coagulation zone geometry created in the tissue: (a) experimental and (b) theoretical coagulation. Scale in ºC.

Fig. 6 shows the depth (D) and width (W) of experimental and theoretical coagulations zones. We always observed significant differences ($p < 0.05$) between groups. The results from theoretical model are in close agreement with the experimental results. Overall, coagulation depth values

computed from theoretical model (Fig. 6 (a)) were within the ranges observed experimentally. However, about coagulation widths (Fig. 6 (b)) there was a slight offset (≈ 2 mm) between theoretical and experimental results. We observed deeper and wider coagulation zones with lower flow rates and higher coolant temperatures, except for the case T1F3.

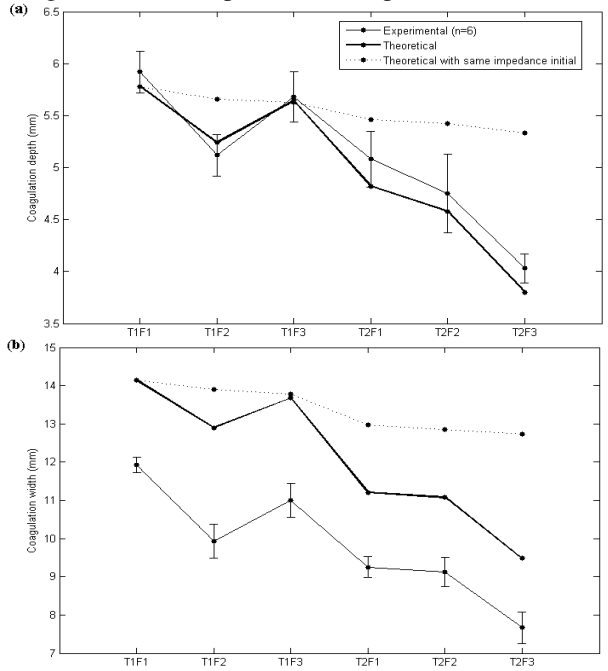

Fig. 6. Comparison between experimental and theoretical coagulation dimensions: depth (a) and width (b). In the experimental results, the dots represent the mean (n=6) and the vertical lines represent ± standard deviation. Six studies cases: two coolant temperatures (T1: 20ºC, T2: 5ºC) and three flow rates (F1: 50 mL/min, F2: 75 mL/min, F3: 100 mL/min).

IV. Discussion

We developed experimental and theoretical models to study the electrical-thermal behaviour of an internally cooled bipolar electrode. We also assessed the effect of flow rate and coolant temperature on the impedance progress during RF heating and thermal coagulation zone dimensions. We compared experimental and theoretical results to evaluate the feasibility of the theoretical model. The theoretical impedance progress matched well with experimental results for any combination of flow rate and coolant temperature, as shown in Fig. 4. We also observed in all cases that the impedance decreased until a specific time and then it remained more or less constant. Computer simulations allowed to verify that this time corresponds to the instant when the temperature in some point of the tissue reached ≈100ºC. This makes sense: the impedance decrease is associated with the exponential increase of the electrical conductivity with the temperature below 100ºC. From this point on, the sudden decrease in electrical conductivity involves stabilization in the impedance progress.

Moreover, theoretical and experimental coagulation dimensions practically coincided for any of the studied cases (Fig. 6). However, we observed slight differences between them. These could be due to the use of a millimeter ruler in measurements, which could be solved by digitizing the photographs with specific software to measure the coagulation dimensions. As mentioned above, we obtained larger coagulations with lower flow rates and higher coolant temperatures. Usually the opposite happens, that is, larger coagulation dimensions are achieved with cooled electrodes, i.e. with higher flow rate and lower temperatures. This could be due to the particular protocol used for delivering RF power: constant voltage and duration. I.e. results (and hence conclusions) could be different by varying voltage or time. In spite of this, our findings suggest that the theoretical model is suitable to reproduce the electrical (impedance progress) and thermal (coagulation zones dimensions) behavior. Moreover, the unexpected anomalous behavior of the group T1F3 was due to the different value of the initial impedance in this group. Complementary simulations (see dashed-lines in Fig. 6) confirm this point. Finally, it is known that temperature isotherm is an unsatisfying method to describe tissue damage, since it depends on both temperature and time of exposure. As we are interested in the zone of tissue coagulation, which is smaller than the region where tissue damage occurs [10], we used in this study an isotherm line to estimate the boundary of the coagulation zone.

V. Conclusion

The experimental and theoretical results were in good agreement, which suggests that the theoretical model could be useful in the study of new designs of cooled bipolar electrodes.

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