Study of the Static and Dynamic Characterization of the Biological Tissue to Obtain the Temperature Estimation in RF Ablation Using Computer Modeling

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Abstract

Radiofrequency ablation has been used to treat some types of cardiac arrhythmias. We have previously proposed an ARMAX model (non structural) to estimate the temperature in the tissue during ablation. Computer modeling has allowed us to study the temperature distribution by means of solving numerically theoretical models based on partial differential equations, which represent physical phenomena. Now, our objective is to consider the biological tissue as a system with an input (applied voltage) and output (tissue temperature), and search for a transfer function between these variables. The final aim is to have a simple model that could estimate the temperature at each point of the tissue. We solved the model using the finite element method and identified the transfer function between the temperature at 4 mm depth and an applied voltage using a 7Fr and 4 mm electrode. We used COMSOL Multiphysics to solve the electro-thermal problem and MATLAB to obtain the transfer function. The results showed that the variation in the electrical conductivity of cardiac tissue affected only the static gain of the system, while the variation in the specific heat produced a change only in the dynamic system response. However, the variation in thermal conductivity modified both the static gain and the dynamic system response. These results are a first step towards the development of a macroscopic model based in physical principles, which would lead to better temperature estimation during ablation.

1. Introduction

Radiofrequency (RF) catheter ablation is currently used to treat some types of cardiac arrhythmias [1]. This technique uses RF current (≈500 kHz) to produce a thermal lesion and hence a tissue necrosis at the target zone of cardiac tissue causing the arrhythmia. The electrical current is delivered in the tissue through an active electrode of small dimension placed at the tip of a percutaneus catheter and a dispersive electrode of large dimension located on the patient's back. Currently, the temperature estimation at several millimeters around the electrode tip is interesting to ablation procedure. We have previously proposed an ARMAX model (non structural) to estimate the temperature in the tissue during ablation [2]. We know that the tissue characteristics (electrical and thermal conductivity, density and specific heat) vary during the ablation procedure. Computer modeling has allowed us to study the temperature distribution by means of solving numerically theoretical models based on partial differential equations, which represent physical phenomena. Therefore, we hypothesis handle the look for a relationship between the tissue temperature and the applied voltage and variations on the tissue characteristics. We considered the biological tissue as a system with an input (applied voltage) and output (tissue temperature), and searched for a transfer function between these variables [3,4]. We solved the model using the Finite Element Method (FEM) and identified the transfer function between the temperature at 4 mm depth and an applied voltage using a 7Fr and 4 mm electrode. We used COMSOL Multiphysics to solve the electro-thermal problem and MATLAB to obtain the transfer function. The final aim is to have a simple model that could estimate the temperature at each point of the tissue. In this work we study the behavior of the tissue temperature evolution at 4 mm depth varying tissue characteristics, which is the first step to obtain a macroscopic model.

2. Methods

2.1. Description of the theoretical model

We considered an active electrode of 7Fr and 4 mm length made of platinum-iridium. This electrode was located perpendicular to the tissue, which implied a rotational symmetry axis and allowed to consider a two-dimensional model. Figure 1 shows the geometry and

dimensions of the theoretical model, which includes an active electrode with a fragment of probe made of polyurethane, a thermistor embedded into the active electrode tip, a section of coating material around the thermistor, and a fragment of cardiac tissue. The dispersive electrode was modeled as an electrical condition on boundaries far from the active electrode (see Fig. 1). We considered an insertion depth of the electrode into the cardiac tissue (P): 1.25 mm [5]. This is a first approximation in order to model different pressures between electrode and tissue. We recorded the temperature evolution at the point T_4, which is a depth in tissue of the 4 mm.

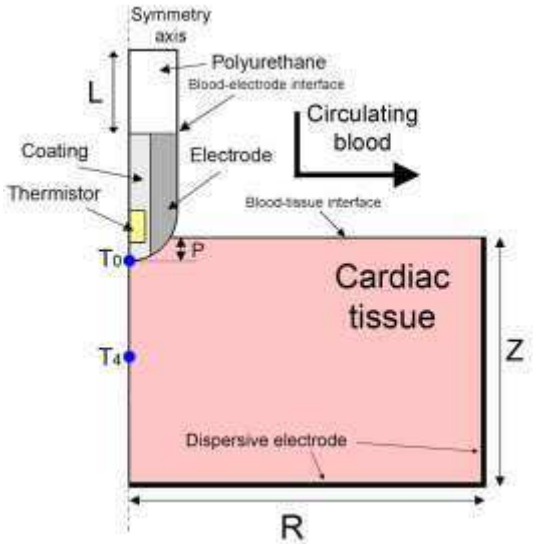

Figure 1. Two-dimensional theoretical model used. The dimensions R=Z=60 mm and L=10 mm are obtained by means of a sensitivity analysis. T_0 is a 0 mm depth and T_4 is a 4 mm depth to evaluated the temperature measure in the biological tissue.

From a mathematical point of view, the model is based on a coupled electric-thermal problem. The thermal problem was solved using the Bioheat Equation [6]:

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

where T is the temperature, t is the time, ρ is the tissue density, c is the specific heat, k is the thermal conductivity, q is the heat source produced by the RF power, Q_p is the heat loss by blood perfusion and Q_m is the metabolic heat generation. The terms Q_p and Q_m are insignificant in RF cardiac ablation and they were not considered [6]. The heat source q (Joule losses) is given by $q = JE$, where J is the current density and E is the electric field strength. The electrical problem was solved using the Laplace equation $\nabla\sigma\nabla V = 0$, where V is the voltage and σ is the electrical conductivity. Electric field is calculated by means of $E = -\nabla V$ and J using Ohm law ($J = \sigma E$). We used a quasi-static approach due to the frequencies used in RF ablation (≈500 kHz) and for the geometric area of interest the tissues can be considered as purely resistive [6]. The blood circulating inside the cardiac chamber was not considered in the model domain. For this reason, the blood-tissue interface and blood-electrode interface were really model boundaries (see Fig. 1).

The electrical boundary conditions were of zero current density in the transversal direction to the symmetry axis, the blood-tissue interface and blood-electrode interface. A boundary condition of voltage zero was set at the dispersive electrode. The voltage at the active electrode tip was of 15 V.

The thermal boundary conditions were of null thermal flux in the transversal direction to the symmetry axis, and of constant temperature at the dispersive electrode and at the outer end of the plastic probe. The circulating blood inside the cardiac chamber produces a cooling effect on the tissue and electrode surfaces. This effect was modeled by means of two thermal convection coefficients h_{tissue} and h_{elec} which represent the cooling effect at the blood-tissue interface and at the blood-electrode interface, respectively. We considered a blood flow rate, which involved a respective values for h_{elec} of 3636 W/m^{2}K and for h_{tissue} of 708 W/m^{2}K [5]. We considered an initial temperature of 37ºC. The values of the physical characteristics of materials were obtained from [4]. In a first approximation we did not consider a change in electrical conductivity of the cardiac tissue of +1.5%/°C and a change in thermal conductivity of the cardiac tissue of +0.001195 K^{-1} [4].

The model was solved using the FEM by means of COMSOL Multiphysics (Stockholm, Sweden). The dimensions R, Z and L (see Fig. 1) were estimated by means of sensitivity analyses in order to avoid boundary effects [4]. In these sensitivity analyses, the value of applied RF voltage was of 15 V. This value was chosen in order to keep maximum temperature lower than 100ºC [6].

2.2. Identification of the system response

In the context of our study the biological tissue was considered to be a dynamic system as [3,4]. This system have a response in terms of transfer functions, i.e. in terms of relationship between the input and output signals. Therefore, the first step was to obtain an accurate identification of the system response of the modeled tissue, which implies to obtain the transfer function G(s). This was done firstly by applying a constant voltage of 15 V during 300 s. The actuator was $u(t)=V^2(t)$, where V was the applied voltage. We checked that temperature stabilized around 55°C at the point T_0. Then, we obtained

the temperature evolution at T_4 depth, which corresponds with a 4mm depth distal of the electrode tip (see Fig. 1). Then, we used the ident command of MATLAB (MathWorks, MA, USA) to estimate the response of the model. This command identifies an arbitrary graphic and returns an estimated form of the transfer function. In our study, this parameter was around 100, which demonstrated a satisfactory identification. We used a transfer function of first-order. The estimated transfer functions G(s) in Laplace-transform quantities had the form:

$$G(s) = \frac{\Delta T_4(s)}{U(s)} = \frac{K_V}{1 + \tau s} \quad (2)$$

where ΔT_4 is the temperature increase (above a temperature base of 37ºC), U(s) is the square of the voltage applied to the electrode, K_V is the steady state gain of the system, expressed in degrees centigrade per square volt, and τ is the time constant.

2.3. Description of the analyzed cases

In order to obtain a structural model based in variations in physical parameters we considered a variations in the tissue characteristics from -75% to 100%, in steps of 25%. In this way, we obtained 8 first order transfer functions corresponding to the changes in the thermal capacity, thermal conductivity and electrical conductivity. We considered these variations to obtain a fan transfer function to study the relationship between the different transfer functions and the different tissue characteristics. The goal is to obtain a macroscopic model of the behavior of the tissue temperature evolution at 4mm depth (T_4).

The assessed parameters were: the dynamic characterization, which was assessed as the time that the tissue takes to reach its final temperature, i.e. associated to the time constant (τ), and the static characterization, which was assesed as the influence the final value of the tissue temperature, i.e. associated to the gain (K_V).

3. Results

3.1. Influence of σ, k and ρc on the first order macroscopic model

Fit for σ variation (electrical conductivity):

Figure 2 shows the evolution of the temperature obtained at the 4 mm depth (T_4) when the electrical conductivity varies. The results show that the electrical conductivity does not affect the time that the tissue takes to reach its final temperature. But, the electrical conductivity influence the final value of the tissue temperature, i.e. higher conductivity leads to higher tissue temperature rise.

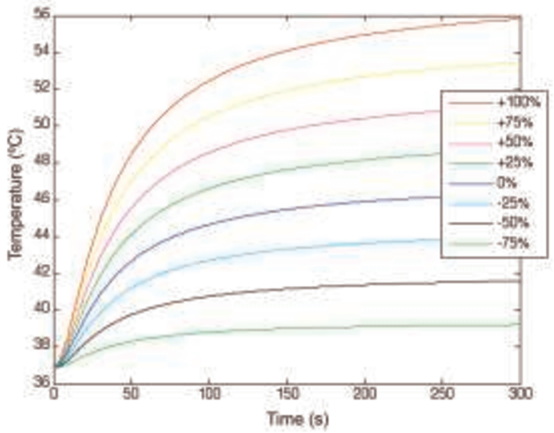

Figure 2. Evolution of the temperature obtained at the 4 mm depth (T_4) when we vary the electrical conductivity.

Fit for k variation (thermal conductivity)

Figure 3 shows the evolution of the temperature obtained at the 4 mm depth (T_4) when the thermal conductivity varies. The results show that the thermal conductivity influences the final temperature obtained. The temperature increase is inversely proportional to the thermal conductivity, i.e.: low thermal conductivity value means higher temperature increase. And the thermal conductivity also influences the time that the tissue takes to reach its final temperature. Again, the dependence is inversely proportional: low thermal conductivity values make the temperature rise slower.

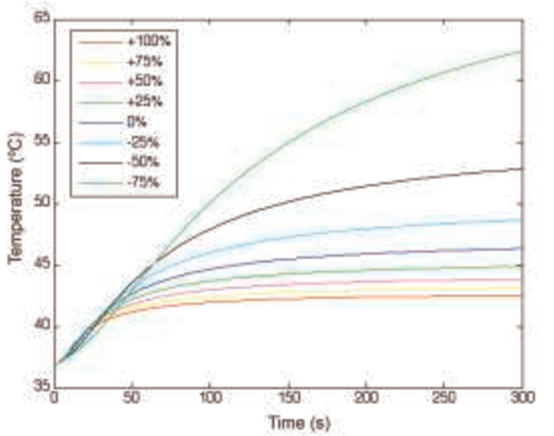

Figure 3. Evolution of the temperature obtained at the 4 mm depth (T_4) when we vary the thermal conductivity.

Fit for ρc variation (thermal capacity)

Figure 4 shows the evolution of the temperature obtained at the 4 mm depth (T_4) when the thermal capacity varies. The results show that the thermal

capacity variation does not affect the final value of the tissue temperature. But, the thermal capacity is directly proportional to the time required to reach the tissue final temperature.

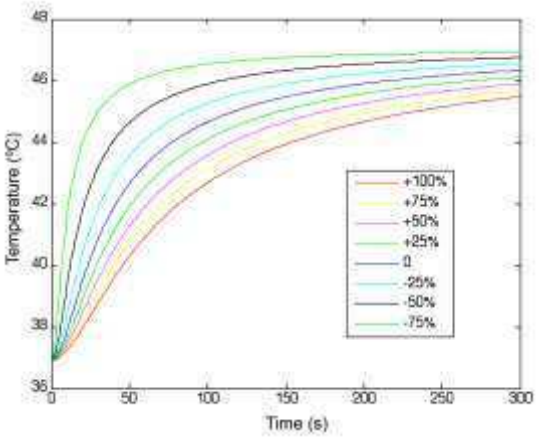

Figure 4. Evolution of the temperature obtained at the 4 mm depth (T_4) when we vary the thermal capacity.

4. Discussion

The objective of this study was to obtain a simple model that could estimate the temperature at each point of the tissue. The firt step was assessed the behavior of the tissue temperature evolution at 4mm depth varying tissue characteristics to obtain a macroscopic model. Regarding the tissue characteristics we observe:

1) The electrical conductivity does not affect the time that the tissue takes to reach its final temperature but, influences the final value of the tissue temperature, i.e. higher conductivity leads to higher tissue temperature rise.

2) The thermal conductivity influences the final temperature obtained and influences the time that the tissue takes to reach its final temperature. The temperature dependence is inversely proportional to the thermal conductivity.

3) The thermal capacity variation does not affect the final value of the tissue temperature but, is directly proportional to the time required to reach the tissue final temperature.

Finally, the unified macroscopic model is possible using the relationship between the time constant in electrical and thermal circuits and on the above conclusions. However, no tests considering the simultaneous variation of electrical conductivity, thermal conductivity and thermal capacity have been performed.

5. Conclusions

The static and dynamic characterization of the biological tissue is a first step to obtain the temperature estimation in RF ablation. We could obtain the development of a macroscopic model based in physical principles, which would lead to better temperature estimation during ablation.

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