# Computer Modeling of Irrigated-tip Electrodes During RF Cardiac Ablation: Comparative Analysis Between Including and Excluding the Problem of Fluid Dynamics

Ana González-Suárez[1], Juan J. Pérez[2], Enrique Berjano[2]

[1]Department of Information and Communication Technologies, Universitat Pompeu Fabra, Barcelona, Spain
[2]Bio-MIT, Electronic Engineering Department, Universitat Politècnica de València, València, Spain

## Abstract

*Accurate modeling of the thermal performance of irrigated-tip electrodes in endocardial radiofrequency ablation requires solving a triple coupled problem involving simultaneous electrical conduction, heat transfer, and fluid dynamics. We propose an approximate method that avoids the fluid dynamics problem by setting a constant value of temperature in a specific zone of the electrode tip. Our objective was to assess the capacity of the approximate method by comparing its results with those obtained from the accurate method. The results showed that the approximate method failed to predict the lesion surface width, overestimating it by 5 mm for any insertion depth and blood flow rate. Likewise, it drastically overestimated the maximum temperature reached in the blood. However, the approximate method is able to predict lesion depth reasonably well (differences lower than 0.1 mm between both methods) and the maximum temperature reached in the tissue (differences always less than 3ºC) throughout the entire ablation time regardless of blood flow rate and electrode insertion depth. In conclusion, the accurate method is the only capable of simulating the real performance of an irrigated electrode-tip surrounded by blood.*

## 1. Introduction

Computer modeling is an analysis technique broadly used in studies on cardiac catheter ablation (RFCA) [1-4]. In order to obtain accurate results, the model has to be as realistic as possible, which implies ever more complicated mathematical formulations, especially in the case of modeling irrigated-tip electrodes for RFCA. In this case an accurate model must be based on a triple coupled problem involving simultaneous electrical conduction, heat transfer, and fluid dynamics [2], which is necessary to model the thermal effect of the circulating blood around the electrode/endocardium and its interaction with the saline infused through the holes at the irrigated electrode tip. Furthermore, the fluid dynamics problem forces the model to be three-dimensional, with the consequent additional computational cost. As an alternative, an approximate method aimed at avoiding the fluid dynamics problem that works with a bidimensional model has been used in previous studies [3,4]. Briefly, thermal cooling due to electrode irrigation was modeled by holding a fixed temperature similar to that registered in clinical practice in a zone of the electrode-tip. Although it has been suggested that this approximate method could reproduce lesion depth reasonably well, a direct comparison between the approximate method and the accurate method has never been conducted in terms of lesion dimensions and maximum temperatures reached in the blood and tissue. Our goal was thus to assess the possibilities of the approximate method in predicting thermal lesion dimensions and blood/tissue temperature distributions by comparing its results with those obtained by the accurate method recently developed and validated experimentally by us [2]. It is especially important that the approximate method should accurately predict the lesion surface width and the maximum blood temperature achieved around the electrode-tip, since it is know that these parameters are related to thrombus formation [5,6]. It is also important that the model be able to predict the maximum temperature reached in tissue, since values of around 100ºC are associated with steam pops formation [7].

## 2. Methods

### 2.1. Computational modeling

The model consists of a fragment of cardiac tissue and an open-irrigated electrode (7Fr diameter and 4 mm length) surrounded by circulating blood (see Figure 1). The electrode was placed perpendicular to the cardiac surface and inserted into the tissue to a depth $D_E$, which

 ISSN: 2325-887X DOI:10.22489/CinC.2017.145-400

varied in the simulations from 0.5 to 1.5 mm. The open-irrigated electrode has multi-holes distributed around its entire distal tip, through which a saline solution is continuously flushing and mixing with the circulating blood. In the case of the approximate method, the saline irrigation was modeled by fixing a temperature at the electrode tip [3,4], while in the accurate method this was done using an inlet velocity boundary condition applied to the electrode surface with the irrigation holes [2]. The dispersive electrode was always modeled as an electrical boundary condition at a distance from the active electrode (bottom surface). Cardiac tissue thickness (H) was 20 mm [1,2] and the remaining dimensions of the fragment of cardiac chamber (X, Y and Z) were estimated by means of a convergence test in order to avoid boundary effects.

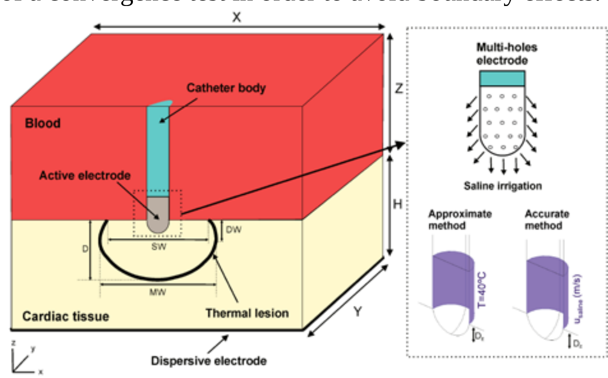


Figure 1. Geometry of the computational model built.

The thermal and electrical properties of the model elements are obtained from [1]. The initial model temperature was 37ºC, except in the electrode tip which was 22ºC [2]. The electrical ($\sigma$) and thermal conductivity ($k$) of cardiac tissue were temperature-dependent piecewise functions defined as in [1].

Electrical and thermal boundary conditions were applied to the model. We modeled a constant power of 10 W for 60 s, which is the usual ablation mode for open-irrigated electrodes in RFCA. We implemented a proportional-integral control algorithm using MATLAB (MathWorks, Natick, MA, USA) and the applied voltage was modulated to maintain the delivering power within 3% of the target [2]. Therefore, a voltage boundary condition was hence applied at the active electrode surface. All the outer surfaces of the model (except the bottom surface) were fixed to a null electric flux. The voltage on the bottom surface was set to 0 V to model the dispersive electrode. For thermal boundary conditions, a null thermal flux was used on the symmetry plane and a constant temperature of 37ºC was fixed on the outer surfaces of the model.

The model is based on a coupled electric-thermal problem, which is solved numerically using the Finite Element Method (FEM) with COMSOL Multiphysics software (COMSOL, Burlington, MA, USA). The governing equation for the thermal problem is the *Bioheat Equation*, modified by the enthalpy method [1] that includes the phase change to model tissue vaporization:

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

where $\rho$ is density (kg m$^{-3}$), $h$ enthalpy, $t$ time (s), $k$ thermal conductivity (W m$^{-1}$ K$^{-1}$), $T$ temperature (ºC), and $q$ the heat source caused by RF power (W m$^{-3}$). In a more general form, model (1) could also include terms accounting for heat loss caused by blood perfusion and for metabolic heat generation. However, since it has been demonstrated that the blood flow away from the coronary arteries does not have significant influence on the temperature distribution during RFCA, and the metabolic heat generation is negligible in comparison to the other terms [8], such quantities are not considered here.

At RF frequencies (≈500 kHz) and over the distance of interest, the biological medium can be considered almost totally resistive, and a quasi-static approach can therefore be used to solve the electrical problem, which allows calculating the value of $q$ [8].

## 2.2. Modeling the blood-saline interaction

The effect of blood circulating inside the cardiac chamber and the saline irrigation were modeled by an approximate or accurate method (see below). We compared the thermal lesion dimensions and the maximum temperature values reached in the tissue and blood computed by the approximate method with those obtained by the accurate method. The thermal lesion shape was assessed by the 50ºC isotherm and was characterized using the following values (see Figure 1): maximum depth (D), maximum width (MW), depth at the maximum width (DW), and surface width (SW).

<u>Approximate method</u>

The effect of blood circulating inside the cardiac chamber was modeled by thermal convection coefficients at the electrode–blood ($h_E$) and the tissue–blood ($h_T$) interfaces, considering electrical conductivity of blood independent of temperature (as in Method 2 described by [1]). Each coefficient was calculated under conditions of high and low blood velocity flow as in [1]. The effect of saline irrigation through the holes at the electrode tip was modeled by fixing a constant temperature of 40ºC only in the cylindrical zone of the electrode tip (see Figure 1), leaving the semispherical tip inserted into the tissue free, as in previous computational studies [3,4]. This temperature value was chosen due to its similarity with that obtained with multi-hole electrodes in clinical practice [9].

<u>Accurate method</u>

The effect of the interaction of blood motion and saline flow inside the cardiac chamber was modeled by including fluid dynamics equations coupled with the heat transfer equation as in [2].

Velocity boundary conditions were applied in this method to model the interaction of blood motion and saline flow. A no slip condition was applied on the upper surfaces of the fluid volume and at the tissue-blood and electrode-blood interfaces. An inlet velocity boundary condition was applied to the left surface of the fluid volume to simulate the two blood velocities of 8.5 and 3 cm/s for high and low flow rate, respectively. An outlet boundary condition of zero pressure was fixed on the right surface of the fluid volume. The saline irrigation flow was taken into account by an inlet velocity condition in the blood region, applied to a specific part of the electrode-blood interface surface where the holes were located, except in the part of the electrode tip inserted in the tissue (see Figure 1).

## 3. Results

Figure 2 shows the progress of lesion depth, surface width and maximum width for both methods. The results show that the approximate method fails to predict the lesion surface width, overestimating it by 5 mm in comparison with the accurate method for any insertion depth, and especially at low blood flow rate. Above all, it was noticeable that the surface lesion width is really created some seconds after starting the RF ablation (see the accurate method), while the approximate method predicts surface width incorrectly, since the lesion is created almost from the beginning of RF ablation. In contrast, lesion depth is correctly predicted by the approximate method, regardless of the blood flow rate and the electrode insertion depth (differences lower than 0.1 mm between both methods) and that this predictive capacity is maintained throughout the entire ablation time.

Figure 3 shows the progress of the maximum temperatures reached in the tissue and the blood for both methods considered. The results show that the approximate method fails to predict the maximum temperature reached in the blood, overestimating it drastically in all cases (differences between both methods were always higher than 15ºC). While the accurate method predicts that blood temperature will remain below 80ºC (which is really the aim of the irrigated-tip electrodes to prevent thrombus formation on the electrode surface), the approximate method wrongly predicts increases of up to 100ºC.On the other hand, the approximate method predicts the maximum temperature reached in the tissue reasonably well, regardless of blood flow rate and electrode insertion depth (differences always lower than 3ºC). Something similar happens with the lesion depth, as this capacity is maintained throughout the entire ablation time.

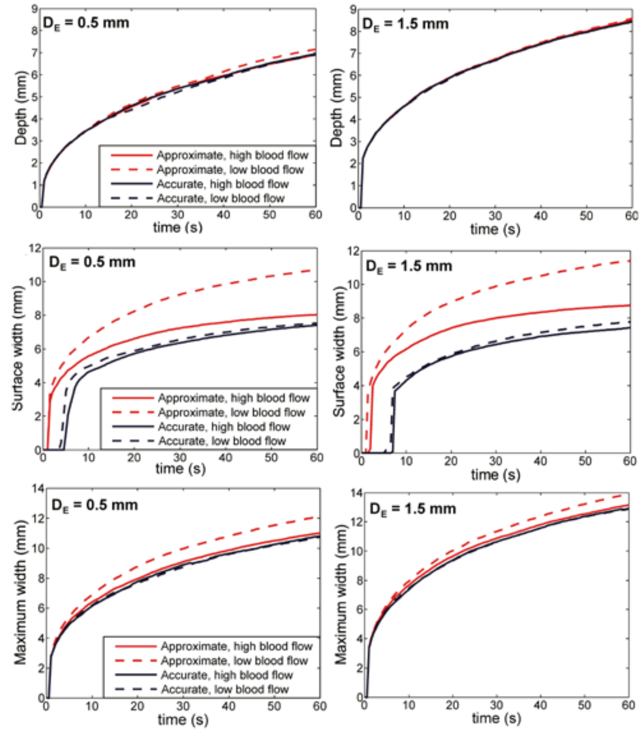


Figure 2. Comparison of the progress of lesion depth, surface width and maximum width between the approximate (red lines) and the accurate (blue lines) methods, for high and low blood flow rates and electrode insertion depths ($D_E$) of 0.5 and 1.5 mm.

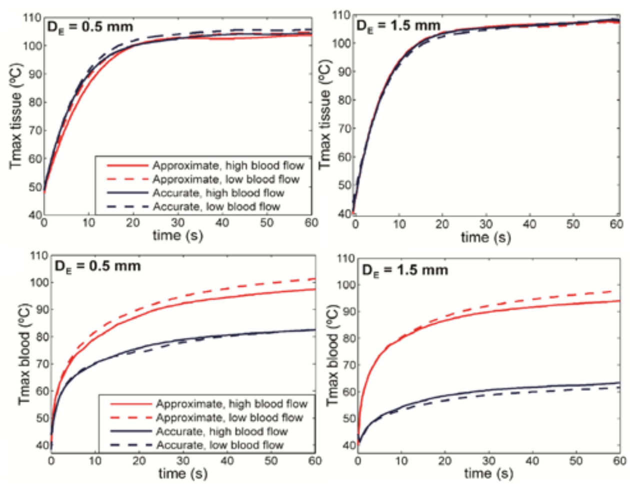


Figure 3. Comparison of the progress of maximum temperatures reached in the tissue and blood between the approximate (red lines) and accurate (blue lines) methods, for high and low blood flow rates and electrode insertion depths ($D_E$) of 0.5 and 1.5 mm.

## 4. Discussion

This computer modeling study assessed the capabilities of the approximate method for modeling the performance of an RF irrigated electrode surrounded by circulating blood (endocardial approach) in terms of adequately predicting thermal lesion dimensions and maximum temperature reached in tissue and blood in comparison with the results computed by the accurate method previously validated by experimental data [2]. The approximate method would be promising whether it could be used instead of the accurate method as it greatly simplifies certain issues such as geometry (often allowing a two-dimensional model), mathematical formulation (reducing the amount of governing equations and boundary conditions) and consequently computational cost.

Our results showed that the approximate method is not suitable for studying temperature distributions in the blood (the blood temperature in the proximity of the electrode-tissue interface was over-estimated reaching a value of ~100ºC in all the cases) nor lesion surface width, which means it is unsuitable for studying factors involved in thrombi formation.

On the other hand, our results confirmed that the approximate method is able to accurately predict lesion depth at all times during ablation, and also the maximum temperature reached in the tissue. This last point is important in terms of computational studies of the circumstances in which a steam pop occurs, since this phenomenon is associated with intratissue temperatures of around 100ºC. In practical terms, a modeling study focusing on issues related with overheating occurring in the tissue (e.g. steam pops) could benefit from these results, since they show the reliability of the approximate method. Although the approximate results were obtained with a three-dimensional model, the results would be the same with a two-dimensional model, further reducing the computational cost.

## 5. Conclusions

The findings confirm that the approximate method is not suitable for predicting either temperature distributions in the blood or surface lesion width, which disables it as a tool to study the factors involved in thrombi formation. Therefore, the accurate method, which solves electrical conduction, heat transfer, and fluid dynamics simultaneously, must be employed for simulating the real performance of an RF irrigated electrode surrounded by circulating blood.


## Acknowledgements

This work was supported by the Spanish "Plan Estatal de Investigación, Desarrollo e Innovación Orientada a los Retos de la Sociedad" under Grant TEC2014–52383-C3-R (TEC2014–52383-C3-1-R and TEC2014–52383-C3-2-R). A. González-Suárez has a Postdoctoral Grant "Juan de la Cierva-formación" (FJCI-2015-27202) supported by the Spanish Ministerio de Economía, Industria y Competitividad.


## References


[1] González-Suárez A, Berjano E. Comparative analysis of different methods of modeling the thermal effect of circulating blood flow during RF cardiac ablation. IEEE Trans Biomed Eng 2016;63(2):250-9.

[2] González-Suárez A, Berjano E, Guerra JM, Gerardo-Giorda L. Computational modeling of open-irrigated electrodes for radiofrequency cardiac ablation including blood motion-saline flow interaction. PLoS One 2016;11(3):e0150356.

[3] Pérez JJ, D'Avila A, Aryana A, Berjano E. Electrical and thermal effects of esophageal temperature probes on radiofrequency catheter ablation of atrial fibrillation: results from a computational modeling study. J Cardiovasc Electrophysiol 2015;26(5):556-64.

[4] Pérez JJ, D'Avila A, Aryana A, Trujillo M, Berjano E. Can fat deposition after myocardial infarction alter the performance of RF catheter ablation of scar-related ventricular tachycardia?: Results from a computer modeling study. J Cardiovasc Electrophysiol 2016; 27(8):947-52.

[5] Matsudaira K, Nakagawa H, Wittkampf FH, Yamanashi WS, Imai S, Pitha JV, et al. High incidence of thrombus formation without impedance rise during radiofrequency ablation using electrode temperature control. Pacing Clin Electrophysiol 2003; 26(5):1227–37.

[6] Yokoyama K, Nakagawa H, Wittkampf FH, Pitha JV, Lazzara R, Jackman WM. Comparison of electrode cooling between internal and open irrigation in radiofrequency ablation lesion depth and incidence of thrombus and steam pop. Circulation 2006;113(1):11–9.

[7] Wood MA, Shaffer KM, Ellenbogen AL, Ownby ED. Microbubbles during radiofrequency catheter ablation: composition and formation. Heart Rhythm. 2005;2(4):397-403.

[8] Berjano EJ. Theoretical modeling for radiofrequency ablation: state-of-the-art and challenges for the future. Biomed Eng Online 2006; 5:24.

[9] Winterfield JR, Jensen J, Gilbert T, Marchlinski F, Natale A, Packer D, Reddy V, Mahapatra S, Wilber DJ. Lesion Size and Safety Comparison Between the Novel Flex Tip on the FlexAbility Ablation Catheter and the Solid Tips on the ThermoCool and ThermoCool SF Ablation Catheters. J Cardiovasc Electrophysiol 2016;27(1):102-9.



Address for correspondence.

Dr. Ana González-Suárez {ana.gonzalez@upf.edu}
Department of Information and Communication Technologies
Universitat Pompeu Fabra
Carrec Roc Boronat 138
08018 Barcelona, Spain