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# **RF-Energized Intracoronary Guidewire to Enhance Bipolar Ablation of the Interventricular Septum: In-silico Feasibility Study**

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## **Abstract**

*Purpose:* Although bipolar radiofrequency (RF) ablation (RFA) is broadly used to eliminate ventricular tachycardias in the interventricular septum wall, it can fail to create transmural lesions in thick ventricular walls. To solve this problem, we explored whether an RF-energized guidewire inserted into the ventricular wall would enhance bipolar RFA in the creation of transmural lesions through the ventricular wall.

*Methods:* We built three-dimensional computational models including two irrigated electrodes placed on opposing sides of the interventricular septum and a metal guidewire inserted into the septum. Computer simulations were conducted to compare the temperature distributions obtained with two ablation modes: bipolar mode (RF power delivered between both irrigated electrode), and time-division multiplexing (TDM) technique, which consists of activating the bipolar mode for 90% of the time and applying RF power between the guidewire and both irrigated electrodes during the remaining time.

*Results:* The TDM technique was the most suitable in terms of creating wider lesions through the entire ventricular wall, avoiding the hour-glass shape of thermal lesions associated with the bipolar mode. This was especially apparent in the case of thick walls (15 mm). Furthermore, the TDM technique was able to create transmural lesions even when the guidewire was displaced from the midplane of the wall.

*Conclusions:* An RF-energized guidewire could enhance bipolar RFA by allowing transmural lesions to be made through thick ventricular walls. However, the safety of this new approach must be assessed in future pre-clinical studies, especially in terms of the risk of stenosis and its clinical impact.

**Key Words:** computer model, guidewire, radiofrequency ablation, septal ventricular tachycardia.

## **Introduction**

In the context of the ablation of resistant ventricular tachycardia (VT), where the reentrant circuits causing the arrhythmia can be located in deep intramural sites, bipolar radiofrequency (RF) ablation (RFA) by two ablation catheters positioned on each side of the interventricular wall has been proposed as an alternative when sequential unipolar application fails to reach intramural circuits [1-3]. However, although bipolar RFA represents progress in the way RF power is used in these cases, it may not be the final answer, as the two ablation catheters are frequently seated more than 15 mm from each other, reducing the potential advantages of bipolar application.

Interestingly, the interventricular wall has septal branches running approximately across the middle of the septum, all which would be less than 10 mm from each side of the septal wall. With the goal of creating thermal lesions deep enough to reach any target in deep intramural sites, we hypothesized that bipolar RF ablation could be enhanced by placing a metal guidewire (0.014" diameter) inside these branches to potentially serve as an RF electrode, i.e. it would become RF-energized. The idea here would hence be to treat the origin of the intramural VT directly, next to the coronary vessels, just where the inaccessibility of the epicardial/endocardial approaches the limit of successful RF ablation [4].

The idea of applying RF power through a 0.035" endovascular RF wire (PowerWire,

Baylis Medical, Montreal, Canada) placed in the interventricular septum has already been clinically proven using 10 W of power for 4 seconds in a monopolar mode [5]. It is known that the application of RF power through a metal wire buried in the tissue has a risk of overheating, steam popping and carbonization due to the high current density at the wire-tissue interface [6]. However, if the applied RF power is correctly chosen, it is possible to create optimal and safe thermal lesions. In any case, the created thermal lesions are generally very small and completely circumscribed around the wire.

Keeping these characteristics in mind, in order to achieve a transmural thermal lesion, i.e. occupying the entire ventricular wall, we proposed that an RF-energized intracoronary guidewire could be employed along with two irrigated-tip electrodes positioned on each side of the interventricular wall. As shown in Figure 1, the guidewire inserted into a septal coronary branch at approximately mid-distance from the two irrigated-tip electrodes. We also proposed to combine sequentially two modes of applying RF power: 1) bipolar mode: between both conventional electrodes while the metallic segment of the guide-wire is not RF-energized; and 2) guide mode: between the guidewire (which would act as an active electrode) and the conventional electrodes (which would serve as dispersive electrodes).

Specifically, we proposed the use of a time-division multiplexing (TDM) technique consisting of activating the bipolar mode for a fraction of the time and an RF-energized guidewire for the remaining time of the repetition period. Our goal was to study the efficacy of this technique by computer simulations in terms of creating transmural lesions on the septal wall and then compare it with the bipolar mode.

## **Materials and methods**

### *Description of the physical situation and computational model*

Figure 1 shows the physical study set-up, which consists of an interventricular septum ablation using two irrigated-tip electrodes placed on opposing sides of the septal wall surface and an endovascular metallic guidewire (0.35 mm diameter) placed inside the septal wall. Both irrigated-tip electrodes and the guidewire could be connected to a programmable switch controller to allow selecting ‘bipolar’, ‘guide’ and ‘time-division multiplexing’ (TDM) ablation modes. The bipolar mode consists of applying energy between both irrigated-tip electrodes, one of which is active and the other dispersive, whereas the guidewire remains passive (non RF-energized). The guide mode consists of applying RF energy only, using the metal zone of the guidewire, while both irrigated-tip electrodes act as dispersive electrodes. Finally, TDM mode consists of combining bipolar and guide mode once a second. In TDM mode, the bipolar pattern is activated for a fraction of the time (90%) and the RF-energized guidewire for the remainder (10%) of each second. In this way the switching frequency between patterns is 1 Hz, i.e. in a given second the bipolar ablation is conducted for the first 900 ms and RF-energized guidewire in the remaining 100 ms.

Figure 2 shows the geometry and dimensions of the three-dimensional computational model built, where the XZ-plane is the symmetry plane. The model consists of a ventricular septum with a metal guidewire placed in the middle. Two irrigated electrodes are placed perpendicularly to the septum wall surface (endocardium–endocardium approach) and surrounded by circulating blood (both ventricles). The irrigated electrodes (7Fr in diameter and 3.5 mm long) are inserted into the tissue to a depth equal to one-third of their diameter

( $D_E \sim 0.8$  mm). The guidewire has a metal part of length  $L$ . Ventricular septum thickness ( $S$ ) is initially 12 mm, as most patients with septal VT have some degree of cardiomyopathy, which causes hypertrophy, in which septal thickness ranges between 9 and 12 mm, or even up to 20 mm [7]. The dimensions of ventricles ( $X$ ,  $Y$  and  $Z$ ) were estimated by a convergence test, in which the value of the maximal temperature achieved in the tissue ( $T_{\max}$ ) after 60 s of RF heating was used as the control parameter. We first considered a tentative spatial (i.e. minimum meshing size) and temporal resolution. To determine the appropriate parameters ( $X$ ,  $Y$  and  $Z$ ), we increased their values by equal amounts. When the difference in the  $T_{\max}$  between consecutive simulations was less than 0.5%, the former values were considered to be adequate. We then determined the appropriate spatial and temporal resolution by means of similar convergence tests using the same control parameter as in the previous test. Discretization was spatially heterogeneous: the finest zone was always the electrode-tissue interface, where the largest voltage gradient was produced and hence the maximum value of current density. In the tissue, grid size was increased gradually with distance from the electrode-tissue interface.

### *Governing equations*

The numerical models were based on a coupled electric-thermal problem which was solved numerically using the Finite Element Method (FEM) with ANSYS software (ANSYS, Canonsburg, PA, USA). The governing equation for the thermal problem was the Bioheat Equation [8]:

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

where  $\rho$  was density ( $\text{kg/m}^3$ ),  $c$  specific heat ( $\text{J/kg}\cdot\text{K}$ ),  $T$  temperature ( $^\circ\text{C}$ ),  $t$  time (s),  $k$  thermal conductivity ( $\text{W/m}\cdot\text{K}$ ),  $q$  the heat source caused by RF power ( $\text{W/m}^3$ ),  $Q_p$  the heat loss caused by blood perfusion ( $\text{W/m}^3$ ) and  $Q_m$  the metabolic heat generation ( $\text{W/m}^3$ ).  $Q_p$  was not considered since its effect is expected to be minor [9] and independent of the method used to model the saline irrigation and blood flow. Likewise,  $Q_m$  was not considered because its effect is negligible in comparison to the other terms [10].

At the frequencies used in RF heating ( $\approx 500$  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 [11]. The distributed heat source  $q$  is then given by  $q = \sigma|\mathbf{E}|^2$ , where  $|\mathbf{E}|$  is the magnitude of the vector electric field ( $\text{V/m}$ ) and  $\sigma$  the electrical conductivity ( $\text{S/m}$ ).  $\mathbf{E} = -\nabla\Phi$  is calculated from the gradient of the voltage  $\Phi$  (V), which, in absence of internal electric sources, satisfies  $\nabla\cdot(\sigma\nabla\Phi) = 0$ .

#### *Model properties and boundary conditions*

The thermal and electrical properties of the model elements are shown in Table 1 [12]. The initial temperature of the entire model was  $37^\circ\text{C}$ . The electrical conductivity ( $\sigma$ ) of cardiac tissue was considered as a temperature-dependent function as follows: it rose linearly  $+1.6\%/^\circ\text{C}$  up to  $100^\circ\text{C}$ , where  $0.6$  S/m was the value of the electrical conductivity assessed at  $37^\circ\text{C}$  (see Table 1), and then it decreased 2 orders for five degrees to model the tissue desiccation process [13].

Figure 3 shows the thermal and electrical boundary conditions. For the thermal boundary conditions (Fig. 3a), a null thermal flux was used on the symmetry axis and a constant



temperature of 37°C was fixed on the outer surfaces of the model at a distance from the ablating electrodes (this was also the initial temperature value). The effect of blood circulating inside the ventricles 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 [15]). These coefficients were calculated under conditions of high blood flow, this being  $h_E = 3346 \text{ W/m}^2\cdot\text{K}$  and  $h_T = 610 \text{ W/m}^2\cdot\text{K}$  [14]. The irrigated electrode was modeled by fixing a temperature of 40°C (as in clinical practice for irrigated electrodes [15]) only in the cylindrical zone of the electrode tip, leaving the semispherical tip free, as in previous computational studies [9,12,16]. This approximation for modeling an irrigated electrode is able to predict lesion depth and also the maximum temperature reached in the tissue at all times during ablation [17].

Regarding the electrical boundary conditions (Fig. 3b-c), all the outer surfaces of the model and the symmetry plane were fixed to have zero electric flux (Neumann boundary condition). As in clinical practice with irrigated electrodes, a constant power ablation mode was used. The electrical current was adjusted by means of calculating the electrical impedance in each time-step in order to maintain the power at a constant value. Knowing the applied current and measuring the applied voltage at the ablating electrode, it is possible to calculate the impedance in each time-step. For the bipolar mode (Fig. 3b), the power was applied at one of the electrodes and another was set to 0 V (dispersive electrode), so the electrical currents were forced to flow between both electrodes. Regarding the guide mode (Fig. 3c), the power was applied through the guidewire and both electrodes were set to 0 V

(dispersive electrodes). In this case, the electrical currents were forced to flow between the guidewire and both electrodes. And the TDM mode was a combination of both ablation modes: for 90% of the time the power was applied in bipolar mode and the remaining time (10%) in guide mode per second.

### *Output variables*

We studied the efficacy of the three different ablation modes in terms of creating transmural lesions through the septal wall after the ablation time (60 s). For this purpose, we conducted a series of simulations to study the effect on the temperature distributions and the thermal lesions created in the septal wall by changing the applied power, septal wall thickness, the size of the metallic guidewire, the guide mode (passive or active) and the displacement of the guidewire from the midpoint inside the septal wall (longitudinal or lateral displacement). Computer simulations were stopped when the maximal temperature in the tissue reached 100°C. The thermal lesion shapes were identified by the 50°C isotherm contour, as in previous studies [9,12,14,16,17], which is usually considered to reasonably represent the isotherm of irreversible myocardial injury in hyperthermic ablation [18].

## **Results**

The convergence test provided the following optimal dimensions:  $X = (80 - S)/2$  mm,  $Y = 22.5$  mm and  $Z = 50$  mm. The optimal grid size was of  $\sim 80$   $\mu\text{m}$  in the finest zone (electrode-tissue interface and guidewire). The model had around 610,940 elements and 110,603 nodes. The time step was 0.2 s for bipolar and guide modes, and 0.1 s for the TDM

mode.

Figure 4 shows the temperature distributions and thermal lesion contour (black line) obtained in a 12-mm thickness ventricular septum after a bipolar RF ablation conducted exclusively with two ablation catheters placed on each endocardial side for three different power levels. The use of a high level of power (20 W) was associated with rapid overheating (temperature of 100°C was reached at ~25 s), when RF power immediately ceased and lesion geometry became hourglass-like. In contrast, a lower level of power (16 W) allowed RF power to be applied for the entire ablation duration (60 s) and lesion geometry was optimum in terms of lesion width in the middle zone. This lesion shape is expected to be more suitable to thermally destroy the abnormal electric circuit responsible for the VT, since it covers the entire wall beneath the RF electrodes. For this reason, from that point on, this power level was assumed to be optimum and was hence used in all the RF bipolar mode simulations.

The next step was to assess distortions of the temperature distributions obtained in the ventricular septum with bipolar RF ablation due to the presence of a passive guidewire (i.e. non RF-energized). Figure 5 shows the temperature distributions and thermal lesion contours (black line) created in the septum for a 16 W bipolar RF ablation. It was noted that whatever the length of the metal zone of the guidewire, the effect of its presence was almost negligible in terms of temperature distributions. The thermal lesion in the middle zone was slightly extended on the guidewire surface only with a very long metal zone (> 10 mm), probably due to the thermal conductivity of the metal being higher than the tissue. No electrical effect was observed due to the passive presence of the guidewire, i.e. at no time

did the guidewire play the role of a secondary RF electrode.

Figure 6 shows the temperature distributions and thermal lesion contours (black line) obtained when RF power was applied in guide mode, i.e. between the guidewire and the irrigated-tip electrodes (RF-energized guidewire, GAc). Since the contact surface of the irrigated-tip electrodes is considerably larger than the guidewire, what really occurs is that the guidewire acts as an active electrode and the irrigated-tip electrodes as dispersive electrodes. Applied power was changed from 16 W to 3 W. As in the case of guide mode ablation, the use of high power levels was associated with rapid overheating (i.e. the maximum tissue temperature rapidly reaches 100°C), which immediately impeded power application and the lesion became circumscribed to the surroundings of the guidewire. The lesion in these cases was clearly not optimal. Accordingly, a power level of only 4 W seems to be the most promising to create a sufficiently wide thermal lesion in the middle zone of the septal wall without causing overheating.

After confirming that a 4 W power level was suitable for energizing the guidewire, we analyzed the effect of increasing the length of the energized guidewire zone (L) from 2.5 to 15 mm using the same power level on the temperature distributions and thermal lesion shapes (see Figure 7). As shown in Fig. 7, short lengths ( $L \leq 5$  mm) were associated with early overheating and thermal lesions confined to around the wire. A length of 10 mm is probably the best to avoid very early overheating and achieve an appreciable lesion volume around the guidewire.

The results obtained so far suggest that while bipolar RF ablation is best at creating especially wide lesions in both endocardial zones (see Fig. 4 and 5), the RF-energized

guidewire is better at creating spatially located lesions in the middle of the septum (see Figs. 6 and 7). From this analysis, our proposal was to combine both ablation modes by means of a time-division multiplexing (TDM) technique. In order to make TDM feasible a constant power of 16 W was applied throughout the ablation, i.e. the same power level was used during the standard bipolar ablation and the RF-energized guidewire ablation. Figure 8 shows the effect of increasing the length of the guidewire energized zone (L) from 2.5 to 15 mm using TDM on temperature distributions and thermal lesion shapes. Although early overheating still occurred for short wire lengths, the results showed an optimal lesion filling almost all the septum zone beneath the irrigated-tip electrodes using the technique for larger guidewires. The best thermal lesion was achieved with a 10 mm guidewire without overheating the tissue (the maximum temperature reached in the tissue is 89.3°C).

In order to check the possible advantage of TDM over the sole bipolar ablation, Figure 9 compares the temperature distributions and thermal lesion contours (black line) obtained for both modes when septal wall (S) thickness varies between 10 and 15 mm. The TDM technique allowed wider lesions to be created in the middle of the ventricular wall, avoiding the hour-glass shape associated with the bipolar technique, especially with very thick ventricular walls (15 mm, Fig. 9f).

Figure 10 shows the effect of displacing the guidewire to one side of the septal wall on the temperature distributions and thermal lesions. We observed that although the guidewire was not located exactly in the middle of the septum, the thermal lesion was equally effective (i.e. transmural). However, an increment of up to 7°C was observed in the maximum temperature reached in the tissue between the guidewire and nearest irrigated-tip

electrode when the guidewire was displaced 3 mm away from the midplane (Fig. 10d).

In contrast, as shown in Figure 11, the effect of a vertical guidewire displacement along the midplane made the lesion extend along the guidewire itself, while the maximum temperature was slightly reduced (around 3°C when the guidewire shifts up to 10 mm from the midpoint, Fig. 11e).

## **Discussion**

This computer modeling study explored whether a guidewire placed in a coronary vein in the middle of the septum could enhance the RF ablation of ventricular arrhythmias by applying RF power directly between itself and a couple of irrigated electrodes on each side of the septal wall. Since the aim was to improve the current bipolar mode conducted with these electrodes, the first simulations were intended to characterize this mode. The results were similar to those found in the case of bipolar ablation with two internally cooled electrodes [19]: a power of 16 W was optimal to create transmural lesions through the ventricular wall and simultaneously avoid an hourglass-like shape (see Fig. 4c). In fact, bipolar ablation is used for resistant septal VTs and should take preference over the new method based on an RF-energized guidewire. However, bipolar septal ablations not only fail to control septal VTs (by rendering them non-inducible) in approximately 50% of patients, but are also related to a large number of clinical recurrences (similar to ethanol septal injection) [20]. The main reason for bipolar ablation not working in septal VTs is supposed to be the thickness of the septal wall. In clinical practice, bipolar ablation works well when the 2 active catheters are less than 10 mm from each other. The difficult septal

VTs are typically located deep inside the septal wall, where bipolar ablation cannot reach, so that an RF-energized guidewire could be a clinical option for treating this type of VT when alternative methods cannot work.

In this respect, the first idea was to take advantage of the metal in the guidewire to make it act passively as a secondary electrode immersed in the current density field created by the pair of irrigated electrodes. This phenomenon has previously been observed in small metal pieces placed near RF ablation electrodes [21], and hence could be exploited to enlarge the lesion around the guidewire. However, the computer results suggest that the role of the metal part of the guidewire as a secondary electrode is practically negligible (see Fig. 5), and so it has to be made to actively deliver RF power.

If the metal part of the guidewire acts as a primary electrode, it seems obvious that due to its having a smaller area than the irrigated electrodes, these will act as dispersive electrodes. The results confirmed this theory, at least for relatively high powers (14-16 W, see Fig. 6a-b). However, under these conditions, early overheating occurs around the guidewire, which would cause steam pops, as observed experimentally in a previous ex vivo study in which a 0.5 mm guidewire was energized with 20 W for 20 s [6]. Consequently, low power has to be programmed when a thin guidewire is RF-energized in order to avoid rapid overheating and hence to make the thermal lesion grows. Our results suggest a value of 3-4 W (see Fig. 6g and 6h), and a metal guidewire length of ~10 mm (see Fig. 7). Interestingly, this power level is very similar to those employed in experimental RF cardiac ablation studies with needle electrodes (3-4 W for a 0.4 mm electrode [22], and ~2 W for a 0.5 mm electrode [23]). In fact, once inserted into the tissue,

a needle electrode will act as a guidewire from an electrical and thermal point of view.

Summarizing thus far, while the bipolar mode with 16 W allows especially wide thermal lesions in both endocardial zones (see Fig. 4c), a guidewire energized with 3-4 W seems to be suitable for creating lesions spatially located in the middle of the septum (see Fig. 6g-h). In order to create a thermal lesion sufficiently wide through the entire ventricular wall, the time-division multiplexing (TDM) technique is expected to combine the advantages of each mode. Although the advantage of TDM over the bipolar mode is not evident for a 12 mm ventricular wall (Fig. 8c vs. 5e), it does become relevant when a thicker wall is considered. For instance, for a thickness of 15 mm, while bipolar failed to create a wide lesion on the median plane (see Fig. 9c), TDM was able to create an equally broad lesion through the entire ventricular wall (see Fig. 9f). The results also suggest that this advantage would remain when the guidewire is displaced both from the median plane of the wall (see Fig. 10) and with respect to the axis of the two electrodes (see Fig. 11).

As a limitation, the models did not solve the fluid dynamics problem, but they did use thermal convective coefficients at the electrode-blood and tissue-blood interfaces to simulate the thermal effect caused by circulating blood. Although this approximation is not enough to obtain a realistic blood temperature distribution, it can accurately estimate lesion depth in the tissue and also the maximum temperature reached in the tissue [17]. Since we were not interested in the blood temperature distributions (necessary, for example, for studying factors involved in thrombus and char formation on the electrode) we used thermal convective coefficients to reduce the mathematical framework of the problem (avoiding fluid dynamics calculations) and consequently the computational cost.



### *Clinical risks and technical limitations*

This computer modeling study was conducted to assess whether an RF-energized intracoronary guidewire would enhance bipolar ablation of the interventricular septum, especially when using a TDM technique. Although the results are promising in terms of achieving transmural ventricular wall lesions of up to 15 mm thick, the use of an RF-energized guidewire could involve clinical risks.

The creation of a thermal lesion within a vessel will certainly make the vessel wall partially collapse due to RF-induced heating. This phenomenon has been observed in other procedures based on endovascular RF ablation, such as endovenous ablation varicose vein treatment [24] and RF-based renal denervation [25]. As in these procedures, the main concern is avoiding stenosis, which mostly occurs in small vessels [25]. Stenosis could reduce or even completely stop the blood supply in a specific zone of the ventricular wall. Future pre-clinical studies should therefore be planned to assess the safety of the RF-energized intracoronary guidewire in terms of stenosis and its clinical impact. As in the case of other procedures based on endovascular RF ablation, the degree of stenosis could be related to variables such as the endovenous lesion length and applied power level [25].

If the results of future pre-clinical studies reveal that complete stenosis occurs after using the RF-energized guidewire, pre-treatment angiographic mapping of the damaged zone would make it possible to assess if this risk is acceptable. In order to reduce the risks as far as possible, we consider that the RF-energized guidewire should be introduced into a venous rather than an arterial branch, as recently proposed for the ethanol injection [26]. The technical limitations could be related to the metallic guidewire sticking to the inner

wall of the vessel, which has been observed in endovenous varicose vein RF-ablation [27].

Even though there are potential clinical risks associated with the use of RF-energized intracoronary guidewire its clinical safety will only be confirmed after further pre-clinical and clinical studies, possibly focused on the occurrence of RF-induced stenosis. In this respect, preliminary bench tests, such as the in-silico model presented here, play a crucial role in roadmapping the development of a new medical device. In other words, certain issues related to clinical safety (e.g stenosis occurrence and its clinical impact) are clearly beyond of the scope of our computer model.

Additionally, from the procedural point of view, it has to be accepted that it will sometimes be difficult to position the two catheters and the guidewire, especially if only fluoroscopy is employed. However, electroanatomical mapping systems greatly facilitate this task and allow both catheters to be visualized and correctly positioned. The computer results suggest that once both catheters have been accurately placed, slight displacements of the guidewire do not impair the performance of the new technique (see Figs. 10 and 11).

In short, despite the risk of stenosis, we think that it could be acceptable to deliberately deliver RF energy into a septal branch to treat a VT under specific justified clinical situations. As in ethanol ablation, the RF-energized intracoronary guidewire could be considered exclusively as a bail-out technique for cases refractory to bipolar ablation [28].

## **Conclusions**

The use of an RF-energized guidewire combined with the TDM technique could enhance bipolar ablation aimed at eliminating a ventricular tachycardia (VT) in the interventricular

septum wall, especially in walls up to 15 mm thick. The safety of this new approach must be assessed in future pre-clinical studies, especially in terms of the risk of stenosis and its clinical impact.

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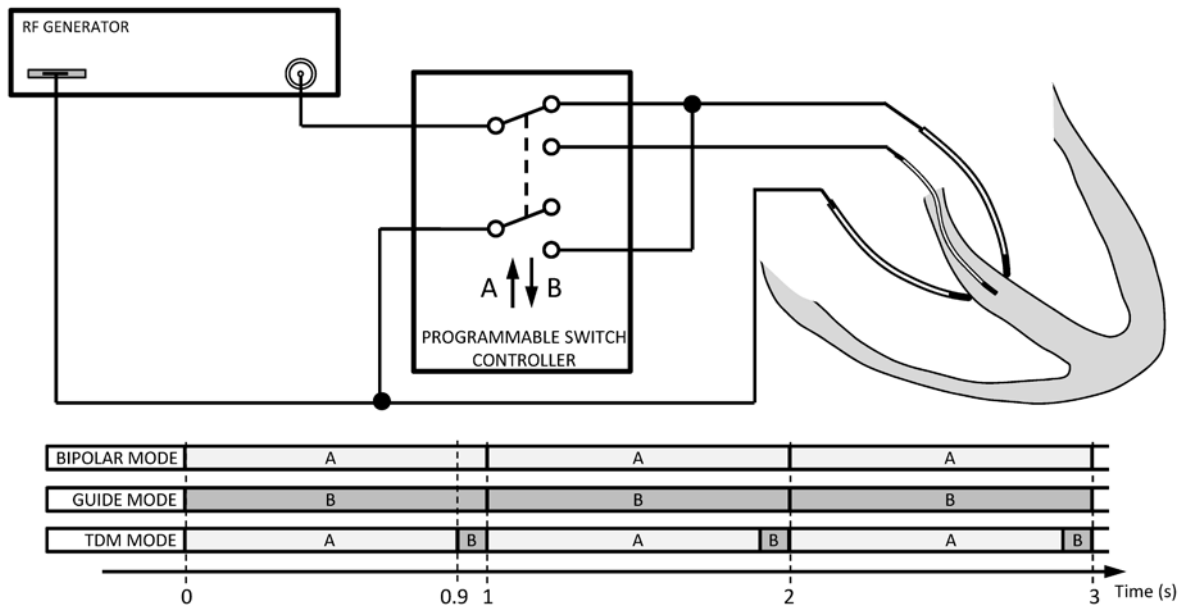
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**Table 1** Physical characteristics of tissues and materials employed in the computational models (data from [12]).

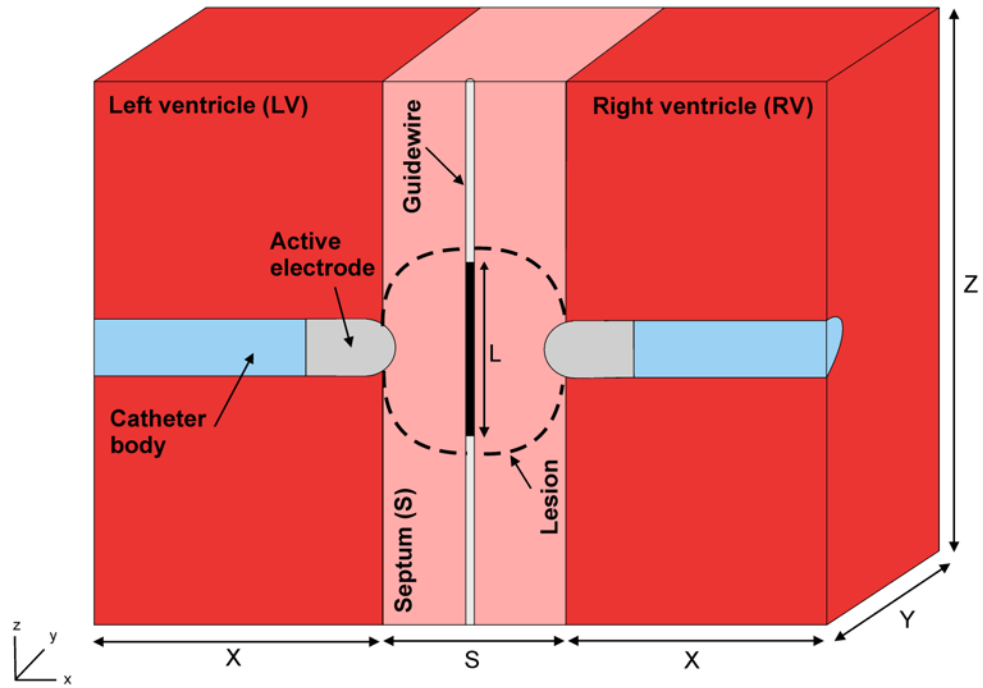
Tissue	$\sigma$ (S/m)	$k$ (W/m·K)	$\rho$ (kg/m <sup>3</sup> )	$c$ (J/kg·K)
Cardiac tissue	0.61	0.70	1200	3200
Cardiac chamber/Blood	0.99	0.54	1000	4180
Electrode, guidewire/Pt-Ir	$4.6 \times 10^6$	71	21500	132
Catheter/Polyurethane	$10^{-5}$	0.026	70	1045

$\sigma$ : electric conductivity;  $k$ : thermal conductivity;  $\rho$ : density;  $c$ : specific heat

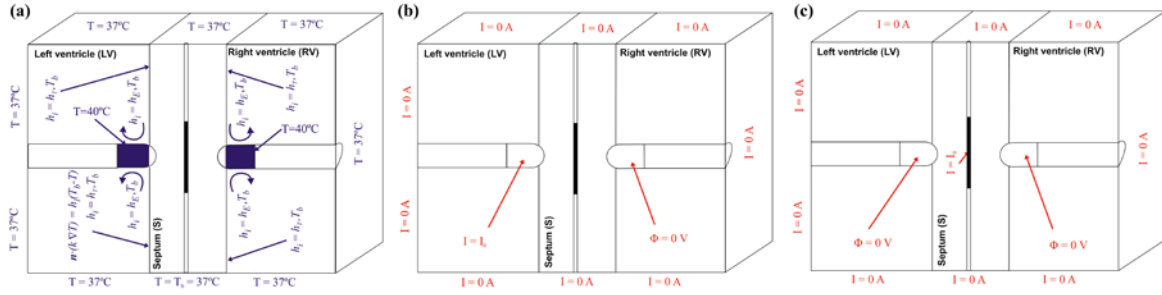


**Figure 1** Physical arrangement for the proposed technique showing two ablation catheters positioned on opposing sides of the interventricular septum and a guidewire inserted into a septal coronary branch. The switching controller can implement the time-division multiplexing (TDM) technique, which consists of activating the bipolar mode for a relatively long fraction of time (A) and RF-energized guidewire in the remaining time (B).

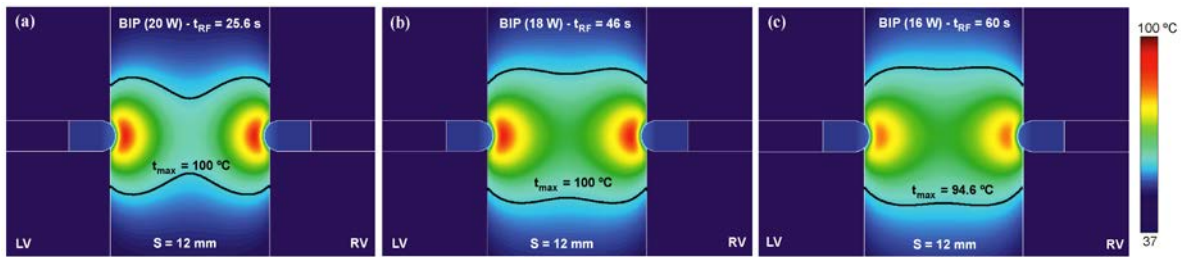




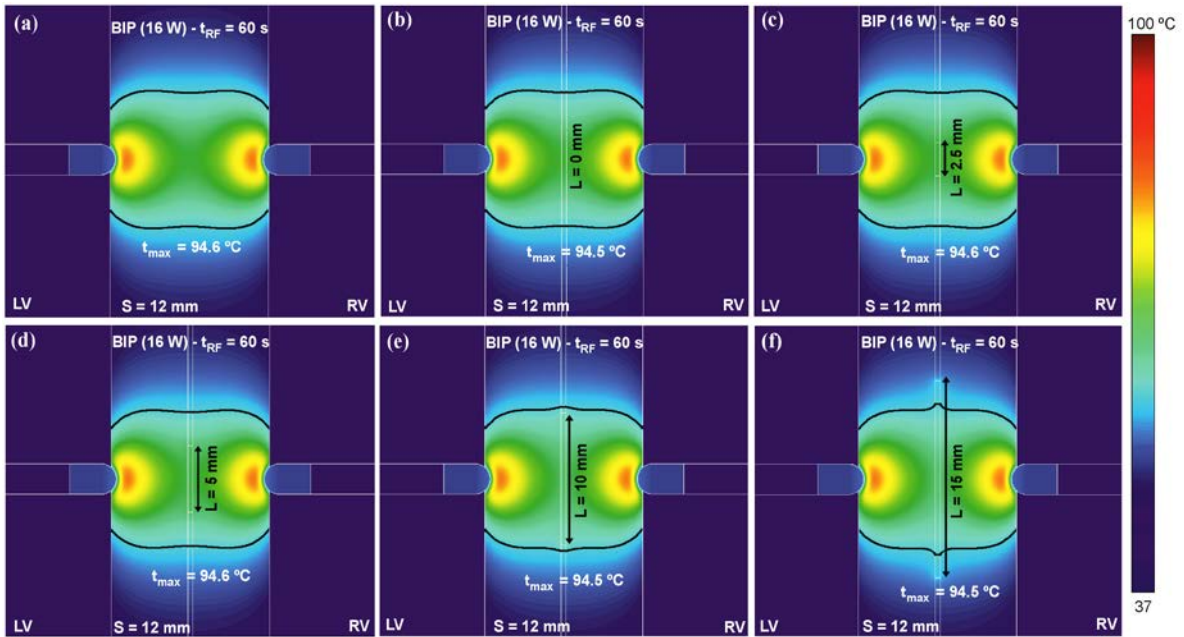
**Figure 2** Geometry of the three-dimensional computational model built (not to scale) consisting of a ventricular septum (thickness  $S$ ) in which a metallic guidewire has been placed. RF current can be programmed to flow either in bipolar mode between a couple of irrigated-tip electrodes (7Fr, 3.5 mm), or between a metal guidewire and the irrigated-tip electrodes. Dimensions of cardiac tissue and blood ( $X$ ,  $Y$  and  $Z$ ) were obtained from a convergence test.



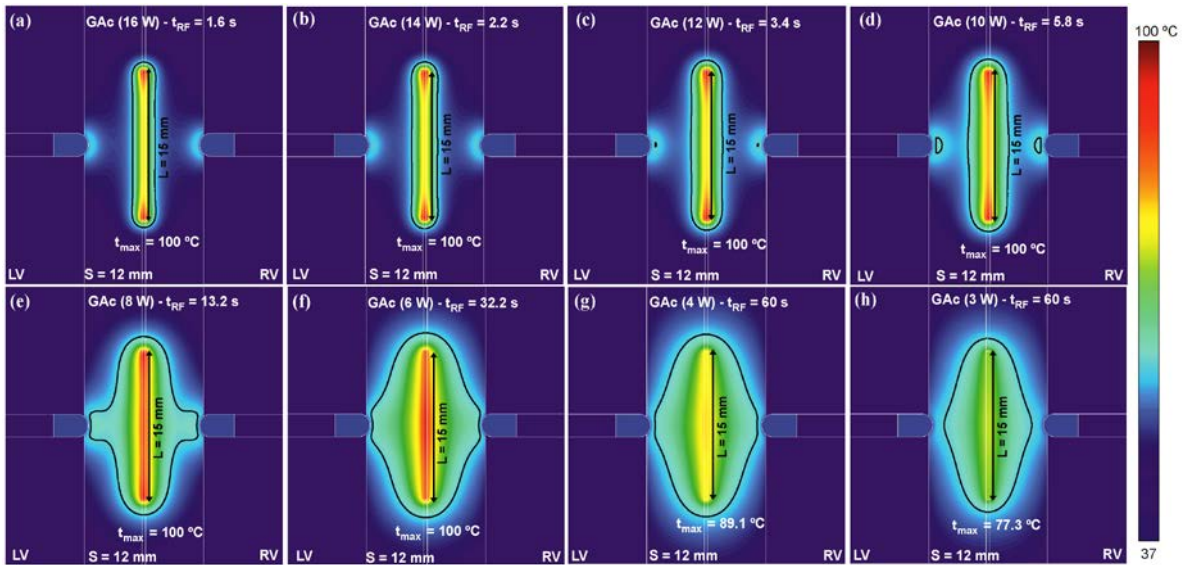
**Figure 3** Thermal (a) boundary conditions of the model.  $h_E$  and  $h_T$  are the thermal convection coefficients at the electrode–blood and the tissue–blood interfaces, respectively. Electrical boundary conditions for the bipolar (b) and guide (c) mode. The guide mode uses the guidewire as active electrode and both irrigated-tip electrodes as dispersive electrodes. The TDM mode is a combination of (b) and (c): RF energy is applied 90% of the time in bipolar mode and the rest of the time (10%) in guide mode.



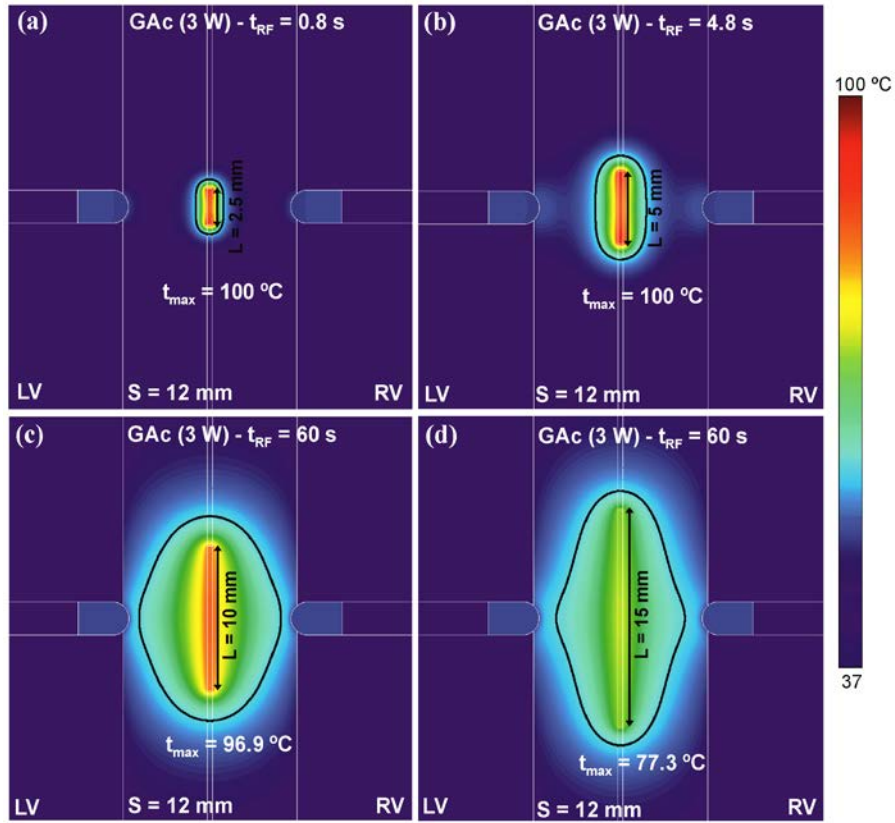
**Figure 4** Temperature distributions and thermal lesion contour (black line) in a 12-mm thickness ventricular septum without guidewire for three values of applied power (20, 18 and 16 W) in bipolar mode.



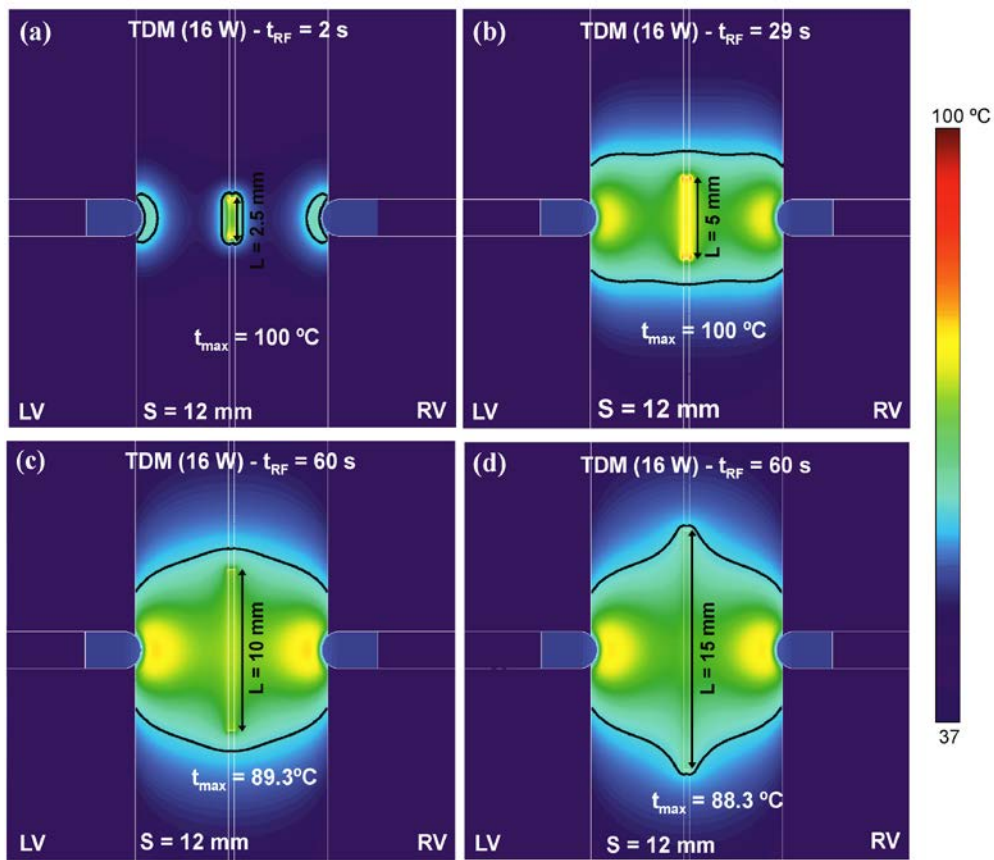
**Figure 5** Effect of the presence of a passive (non RF-energized) metal guidewire placed in the middle of a 12-mm thick ventricular septum ablated in bipolar mode (16 W).



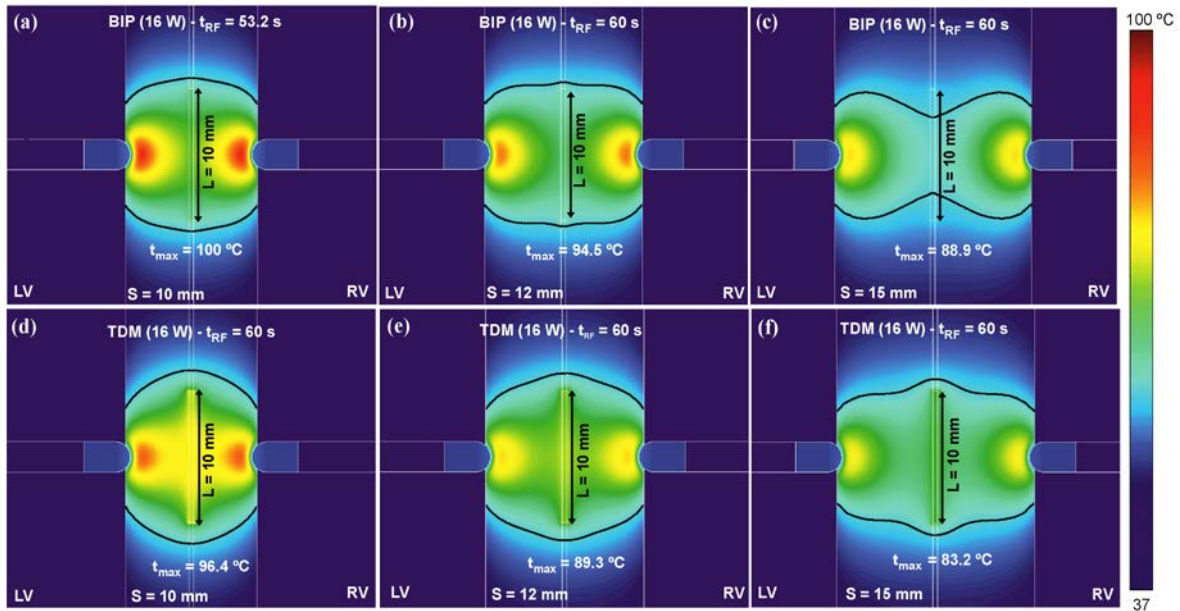
**Figure 6** Temperature distributions and thermal lesion contour (black line) in a 12-mm thick ventricular septum with a RF-energized metal guide wire (15 mm length) for different values of applied power between wire and irrigated-tip electrodes. Computer simulations were stopped when maximal temperature in the tissue reached 100°C ( $t_{RF}$ ).



**Figure 7** Effect of the length of the energized zone in the guidewire (ranging from 2.5 mm to 15 mm) on the temperature distribution in the tissue. Computer simulations were stopped when maximal temperature in the tissue reached 100°C ( $t_{RF}$ ). Power of 4 W was applied between the energized zone and the irrigated-tip electrodes.

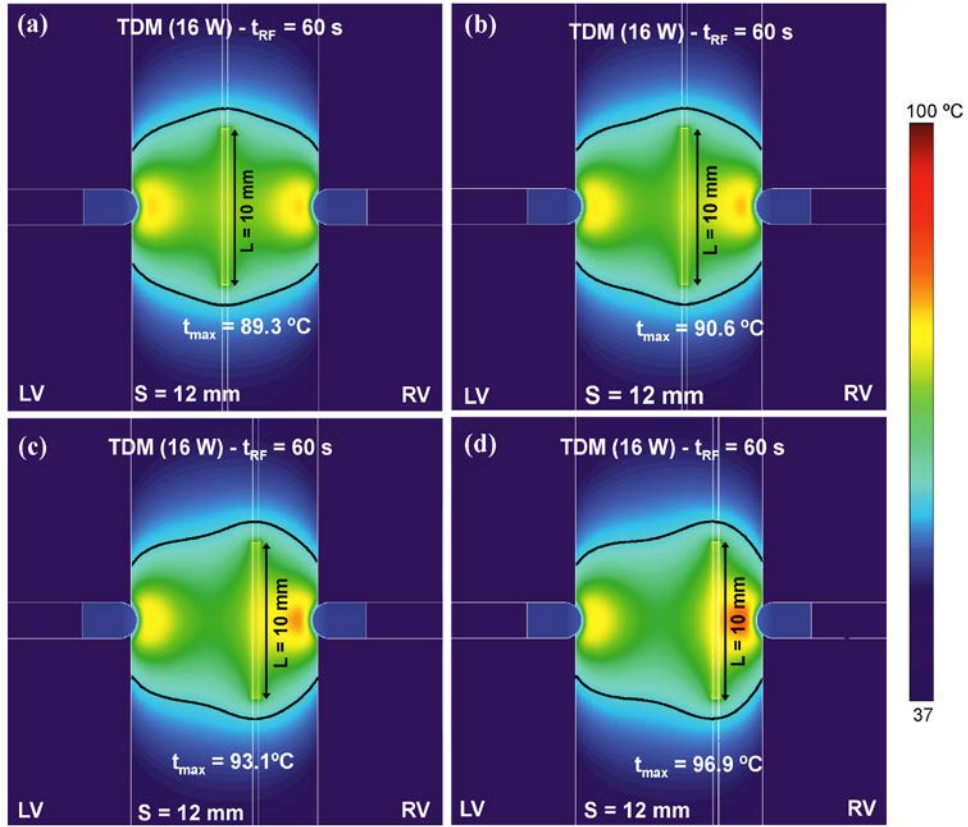


**Figure 8** Temperature distributions and thermal lesion contour (black line) in a 12-mm thick ventricular septum created by time-division multiplexing (TDM) where bipolar RF ablation is conducted for 90% of the time and RF-energized guide wire is conducted 10% of the time (see text for more details). Distributions are those of different lengths of energized zone in the guidewire, from 2.5 to 15 mm. Computer simulations were stopped when maximal temperature in the tissue reached 100°C ( $t_{RF}$ ).

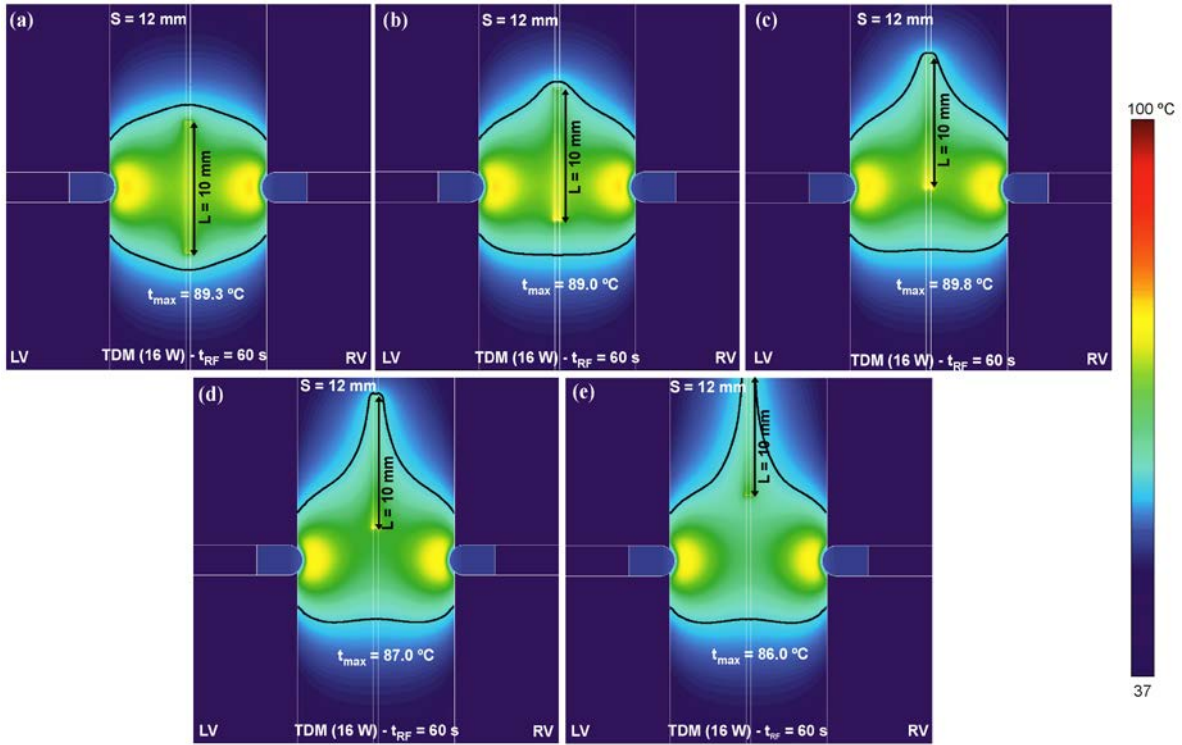


**Figure 9** Comparison between alone bipolar mode (a-c) and time-division multiplexing (TDM) technique (d-f) on temperature distributions and thermal lesion contours (black line) obtained for three thicknesses of ventricular septum (10, 12 and 15 mm). Computer simulations were stopped when maximal temperature in the tissue reached 100°C (t<sub>RF</sub>). A constant power of 16 W was applied throughout the ablation.





**Figure 10** Effect of the displacement of the RF-energized guidewire (10-mm length) to one side on the temperature distributions and thermal lesion contours (black line). The lateral displacement considered from the midpoint of the septum was: (a) 0 mm (without displacement), (b) 1 mm, (c) 2 mm, and (c) 3 mm. Computer simulations were stopped when maximal temperature in the tissue reached 100°C ( $t_{RF}$ ).



**Figure 11** Effect of the displacement of the RF-energized guide wire (10-mm length) along the midplane on the temperature distribution in the tissue and thermal lesion contours (black line). The longitudinal displacement considered from the midplane of the septum was: (a) 0 mm (without displacement), (b) 2.5 mm, (c) 5 mm, (c) 7.5 mm, and (d) 10 mm. Computer simulations were stopped when maximal temperature in the tissue reached 100°C ( $t_{RF}$ ).