A NOVEL ARRAY ANTENNA BASED CATHETER FOR MICROWAVE CARDIAC ABLATION

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Abstract

This paper presents a novel cylindrical slot array antenna based catheter for efficient microwave ablation for the cure of cardiac arrhythmias. The antenna consists of several cylindrical slots fed by a coaxial line to improve the electromagnetic coupling of microwave energy to myocardium. Powerful analysis tools based on FDTD and FEM have been employed for optimizing the antenna array and to predict SAR and temperature distribution. Experiments using fresh ovine heart tissues were performed and the corresponding temperature and lesion sizes were recorded. The predicted temperature profiles agree closely with the measured profiles within the fresh ovine hearts. The temperature profiles of ablations conducted on the ovine tissue indicate that the proposed antenna is capable of producing deeper and narrower lesions.

1. Introduction

Cardiac arrhythmia refers to any conditions where the heart beats at an abnormal rhythm. Over the past years, the radio-frequency (RF) has become a well accepted mode of energy delivery system for cardiac ablation. One of the major disadvantages of RF ablation is that it requires contact between the RF electrode and the cardiac wall to achieve the Joule heating. Since the current RF ablation catheters are designed to produce single discrete lesions, linear lesions utilising RF ablation requires the catheter to be dragged in a linear format. Due to the movement and irregularities of the cardiac walls, maintaining constant contact between the RF electrode and the cardiac wall can be extremely difficult and if there are discontinuities present within the lesion, arrhythmia can re-occur.

Another limitation of the RF ablation is that the lesion created using RF is very shallow and only works on sites where shallow discrete lesions are required.

In contrast to the RF ablation, microwave energy uses dielectric heating by stimulating the oscillation of dipoles in the cardiac tissue and converting electromagnetic energy into heat. Thus, microwave ablation has the potential to ablate tissues at greater depth than RF ablation [2].

In recent years, various types of antennas for microwave catheter ablation have been reported. These include the monopole antenna based catheters[3-4] and cap-choke antenna based catheters[5]. In general, the monopole antennas do not provide lesions of enough depth and produce discontinuous lesions which are not suitable for treatments of many forms of cardiac arrhythmia. The cap-choke antennas produce lesions that are much wider than the depth thus limiting its use in cardiac ablation. Further, cap-choke antenna produces discrete lesions and needs to be dragged inside the myocardium to form a linear lesion, making it unsuitable for certain types of cardiac arrhythmia treatment.

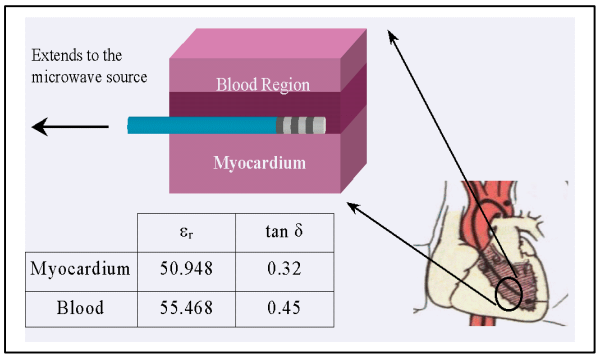
In this paper, we propose a novel slot array antenna based catheter that can produce deep narrow linear lesions. The lesion sizes and, in particular, the depth obtained is sufficient to create transmural atrial ablation. Also, by using different power settings and ablation duration, transmural ventricle ablation can also be achieved.

2. Material and Method

The antenna consists of an array of cylindrical slots fed by a coaxial line. The number of slots, feeding mechanism and the performance of the

antenna have been optimized using FDTD technique and FEM based HFSS™. The HP8510A vector network analyser is used to obtain the permittivity of the myocardium tissue and blood. Fresh ovine hearts and blood are used in this study. The model consists of the antenna surrounded by myocardium and blood. The length of the CSA antenna is 20mm long and the microwave energy is fed to the antenna via a 50 Ω flexible Tflex HF405 coaxial transmission line. The three dimensional computational model is shown in Figure 1. Within the computation model, the myocardium and the blood are differentiated by assigning appropriate measured permittivity for blood and myocardium tissue. The simulations have been performed at 2.45 GHz.

The reflection coefficients of the CSA antenna is measured using the HP8510A vector network analyser. The measured reflection coefficient is compared with the predicted reflection coefficient calculated using the FDTD and the HFSS™. The comparison between measured and predicted reflection coefficient results, as shown in Figure 2, indicate a good agreement between simulation and measurement.



	ε_r	tan δ
Myocardium	50.948	0.32
Blood	55.468	0.45

Figure 1: Computer simulation model.

The computed electric fields are then converted to specific absorption rate (SAR) using the following relationship:

$$SAR = \frac{s}{2}\left|E_{total}\right|^2 \quad \text{W/kg} \qquad (1)$$

The computed SAR values are used to construct the three dimensional temperature distributions within the phantom tissue. The bioheat transfer equation is used to describe the thermal characteristics [7] is given by:

$$rc\frac{\partial T}{\partial t} = \nabla\cdot(k\nabla T) - W_b c_b (T - T_b) + Q_m + Q_{SAR} \quad (2)$$

where ρ is the tissue density (1000 kg/m^3), c is the tissue specific heat (4000 J/kg ^{o}C), k is the tissue thermal conductivity (0.6 W/m ^{o}C), W_b is the blood perfusion, c_b is the blood specific heat, Q_m is the heat generated by the metabolism and Q_{SAR} is the heat generated by the microwave antenna due to the SAR.

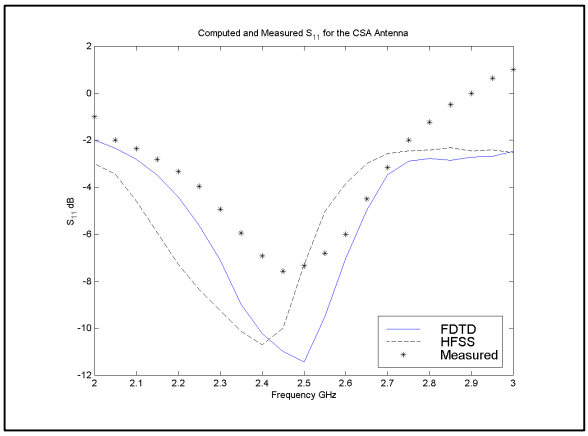


Figure 2: Predicted and measured reflection coefficients.

In this paper, the thermal conductivity, k, is assumed to be constant throughout the computation domain, and the temperature rise due to the metabolism, Q_m, is considered negligible compared to the temperature due to microwave radiation, Q_{SAR}. Also the heat loss due to blood is also not considered for this paper. The simplified equation then becomes:

$$rc\frac{\partial T}{\partial t} = k\nabla^2 T + Q_{SAR} \qquad (3)$$

Equation (3) is solved employing the finite difference scheme on a CM-5 parallel computer using CM Fortran and the thermal patterns in the myocardium tissue have been computed [7]. The finite difference program is used to predict the temperature profile at a depth of 5mm inside the simulated cardiac tissue which is plotted in Figure 3. To confirm the accuracy of the predicted results, measurements have been made on fresh perfused ovine heart tissues closest to the simulated position. The measured results are also plotted in Figure 3 and it can be seen that the predicted temperature profile compares well with the measured results.

A custom-made trolley containing (a) tissue bath, (b) blood flow controller, (c) ambient temperature regulator, (d) microwave system including source, dual directional coupler and power meter, and (e) Luxtron fiber-optic temperature measurement system is used for the

experiment. The schematic of the experiment set up is shown in Figure 4. An aluminum planar shield was used with the trolley to protect the operator from any leakage microwave energy. The tissue bath is filled with either blood or saline phantom solutions. The ambient temperature regulator keeps the temperature of the tissue bath constant at around 37^o to simulate the body temperature during ablation. A roller pump (b) is used to regulate the flow within the tissue bath to simulate the blood flow. The arrow on the tubes indicates the blood flow direction.

A special epoxy grid is made to hold both the tissue sample and the antenna catheter securely within the tissue bath. A coaxial transmission line is clamped on a stand placed behind the tissue bath, with one end of the transmission line connected to the antenna, and the other end connected to the microwave generator.

All in vitro tissue temperature measurements were performed using the Luxtron 3100 thermometry system with four fluoroptic temperature sensors. The temperature probes were placed just below the endocardial surface perpendicular to the microwave antenna. A computer is used to record the temperature distribution before, during and after ablation in real time. It is necessary to measure the temperature before the microwave energy is applied to ensure that the tissue sample has reached equilibrium with the ambient temperature (37^o).

3. Results and Discussion

The experiments were conducted using fresh ovine hearts and blood to test the performance of the CSA antenna at the frequency of 2.45 GHz. Figure 5 shows the temperature profile of the CSA antenna over a 30 seconds ablation period at 100 Watts. It can be seen that for a short period of time, the CSA antenna is capable of raising the temperature over 100 ^{o}C even at a depth of 10mm. In order to lower the temperatures sufficient enough to cause irreversible tissue damage, experiments were conducted at 50W power level at varying duration of 15seconds and 30seconds. For the combinations of 50W, 30 seconds and 50W 15 seconds, the maximum temperatures obtained were 85 ^{o}C and 65 ^{o}C respectively. This shows that both the above mentioned combinations are capable of causing irreversible tissue damage. In Figure 5, results for 100W power exposure are shown to confirm the ablative capability of the CSA antenna. Being able to generate a high temperature gradient at shorter time duration and at lower power levels is one of the key advantages of the CSA antenna, which helps to minimize the operating time on patients. Several lesions have been made using different power and time combinations and the corresponding lesions are digitally photographed and measured. Table 1 shows the lesion size, the surface area and the width to depth ratio.

For cardiac ablation, it is important to keep the width of the lesion as small as possible in order to preserve the healthy cardiac tissues. From Table 1, we can see that using 50 watts and 30 seconds ablation duration, we are able to achieve a width to depth ratio of 0.8 and with 50 watts and 15 seconds we are able to achieve a width to depth ration close to 1. We believe that this is a very significant new result as we not only used low power levels, but the ablation time has also been reduced, while achieving good width to depth ratios. This is critical for ablation in the atrium where healthy cardiac tissue needs to be preserved as much as possible. For thicker tissues, such as the ventricle, a power level of 100 watts can be used. Although at 100W settings, the ratio is not close to 1, it is still very good since there are more muscles present in the ventricle.

Table 1: Lesion sizes for various power/duration combinations.

Power/ duration	Depth (mm)	Width (mm)	Surface area (mm^2)	W:D Ratio
50W 15s	2.9	3.6	14	1.1
50W 30s	4.1	4.3	19.8	0.8
100W 15s	3.9	6.4	27.4	1.6
100W 30s	7.7	10.9	76.9	1.5

4. Conclusion

A novel coaxial slot array antenna that is suitable for microwave ablation in atrium as well as in the ventricle has been proposed and its performance has been extensively analyzed and tested. Overall, the proposed CSA antenna is capable of generating temperatures of greater than 55 degrees required for irreversible lesion formation in a very short period of time. The lesion sizes, in terms of width and depth, are controlled by varying the microwave energy and ablation duration. In relatively short periods of ablation time, the CSA antenna is capable of producing

lesions of sufficient depth for transmural atrial ablation. In some cases, transmural ventricular ablation can also be achieved.

Acknowledgment

This work is supported by the Australian Research Council under the SPIRT grants scheme. The authors thank Mr. Chris McCarthy of Westmead Hospital for his help in fabrication and measurement. The CM5 computer facility is provided by the NSWCPC.

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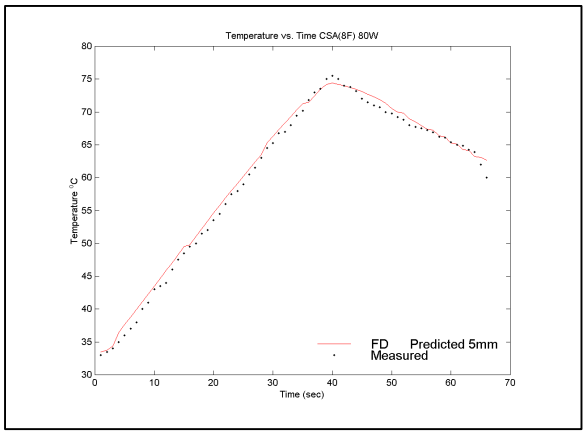

Figure 3: Predicted and measured temperature profile for 60 seconds at 80 W.

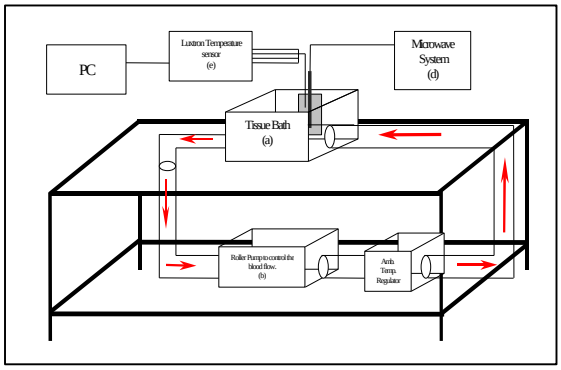

Figure 4. Equipment set up for the experiemnt on a custom made trolley. The equipments are: (a) Tissue bath, (b) Roller pump, (c) Ambient Temperature Regulator, (d) Microwave system and (e) Luxtron optical fiber temperature sensor.

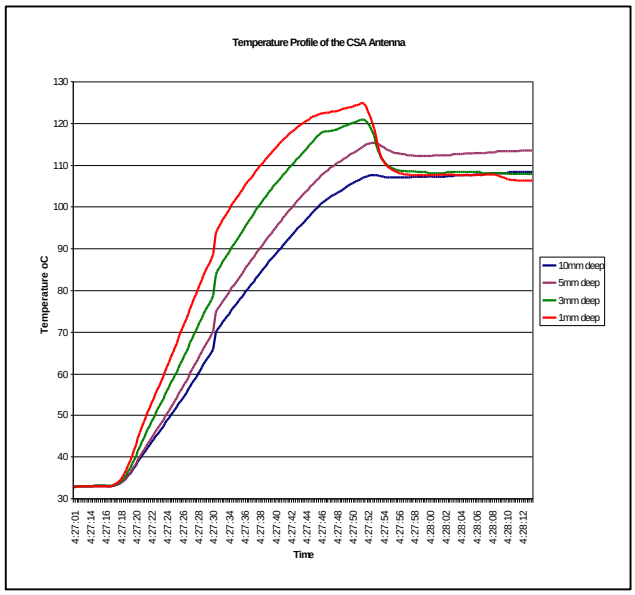

Figure 5: 100 Watt 30 Second duration.

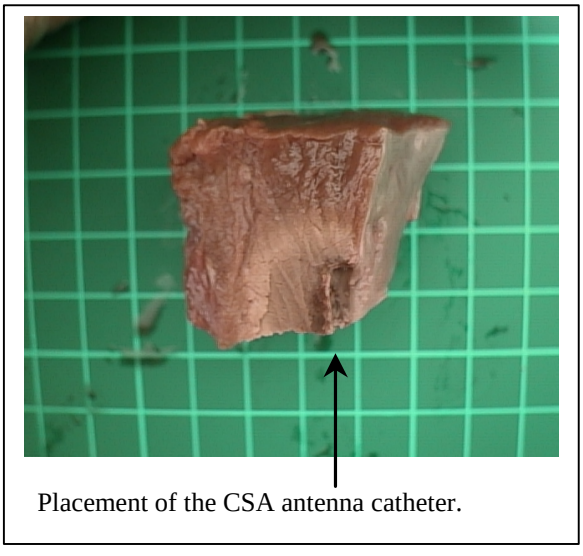

Figure 8: Lesion generated using the CSA antenna. Power = 100 Watts, duration = 30 seconds