Analysis of a Novel Expanded Tip Wire (ETW) Antenna for Microwave Ablation of Cardiac Arrhythmias

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Abstract—A novel expanded tip wire (ETW) catheter antenna is proposed for microwave ablation for the treatment of atrial fibrillation (AF). The antenna is designed as an integral part of coaxial cable so that it can be inserted via a 6F catheter. A numerical model based on the rotationally symmetric finite-difference time-domain technique incorporating the generalized perfectly matched layer as the absorbing boundary condition has been utilized to accurately model the interaction between the antenna and the myocardium. Numerical and in-vitro experimental results are presented for specific absorption rate, return loss and heating pattern produced by the antenna. Both numerical modeling and in-vitro experimentation show that the proposed ETW antenna produces a well-defined electric field distribution that provides continuous long and linear lesions for the treatment of AF.

Index Terms—Atrial fibrillation, cardiac ablation, catheter ablation, finite-difference time-domain, microwave ablation, microwave antenna, specific absorption rate.

I. INTRODUCTION

ATRIAL FIBRILLATION (AF) is one of the most common types of cardiac arrhythmia where the heart beats at an abnormal fast rhythm. Minimally invasive procedures, such as catheter ablation, have become a preferred treatment modality of cardiac arrhythmias in the past decade [1]. These procedures involve the application of electromagnetic energy at various frequencies to the arrhythmogenic tissue by using a delivery device that is introduced to the myocardium via a catheter. The application of electromagnetic energy to an arrhythmogenic site ablates the tissue by raising the local temperature to a point where tissue necrosis is achieved, thus stopping the conduction of undesirable activation signals.

Different ablative strategies are required for the management of different types of cardiac arrhythmia. For the treatment of AF, linear incisions are required to be made in order to direct the undesirable activation signals through a particular path [13]. Radio frequency (RF) catheter ablation has become a popular medical procedure for the treatment of cardiac arrhythmias since the mid-1980s. However, catheter ablation using RF energy produces isolated shallow lesions that are wider than they are deep which makes it difficult to create transmural linear lesions required for the treatment of AF.

To overcome the limitations of the RF ablation, many researchers have investigated the use of alternate energy sources such as microwave energy to ablate the myocardial tissue [1]–[12]. Microwave antennas are the critical elements in the microwave ablation procedure, as the generation of continuous linear transmural lesions depends on the control of radiation characteristics of the antenna. Thus, the ability to create continuous transmural lesions to achieve total isolation of the undesirable activation signals in the atrium is one of the important criteria to judge the efficacy of a particular microwave catheter antenna for the cure of AF.

In general, microwave catheter antennas can broadly be categorized into two types: those antennas that are designed to produce radiation mainly around the antenna tip [5], [8], [10] and those that produce radiation normal to the antenna axis [7]–[10]. Gu et al. [5] recently reported a wide aperture microwave spiral antenna for cardiac ablation which created lesions that are too wide for ablation in the atrium where the available cardiac tissue is limited. The antenna reported by Pisa et al. [9] has shown increased radiation along the antenna length as well as around the tip. The enhanced radiation around the tip of the antenna can be problematic when the antenna is placed near the valves as it may cause unintentional valvular damage due to electromagnetic radiation. Nevels et al. [7] reported several variations of dielectric-loaded and dielectric-coated monopole antennas that provide electromagnetic radiation along the length of the antenna with minimal tip radiation. However, the suitability of those antennas proposed by Nevels et al. [7] cannot be assessed for the treatment of AF due to the lack of experimental results on lesion sizes. Labonte et al. [4] also reported three variations of a monopole antenna, viz., the open tip monopole (OTM), the dielectric tip monopole (DTM), and the metal tip monopole (MTM) antennas. Both the DTM and OTM antennas show excessively high specific absorption rate (SAR) around the antenna junction that can cause nonuniform heating of cardiac tissues. The MTM antenna showed a more evenly distributed SAR, however, it is very poorly matched at 2.45 GHz. The poor impedance matching

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of the MTM antenna results in increased reflected power that can cause unnecessary coaxial cable heating. In order to overcome the above mentioned limitations, we propose in this paper, a novel expanded tip wire (ETW) catheter antenna that is well matched to the source impedance and can create linear transmural lesion for the cure of atrial fibrillation (AF).

The antenna impedance matching and the near-field distribution are important for the success of microwave cardiac ablation. In addition, modeling of antenna-myocardium interaction is useful for the design of an antenna that can couple energy efficiently into the myocardium for creating localized ablation. In this paper, we provide a detailed account of modeling of antenna-myocardium interaction using an efficient rotationally symmetric finite-difference time-domain (FDTD) technique with generalized perfectly matched layer (GPML) [14], [16] absorbing boundary condition (ABC) and validate the calculated results with measurements made on chemical phantom and in-vitro live bovine myocardial tissues.

We employ the RS-FDTD technique with GPML ABC to empirically optimize the ETW antenna for obtaining the optimum dimensions to achieve the best performance of the antenna with an objective to reduce the return loss and to confine the electric-field distribution around the axis of the antenna for maximum energy deposition into the myocardium. Both the computed and in-vitro experimental results show that the proposed ETW antenna produces a well-confined localized heating of the myocardium tissue and is very well suited for creating linear transmural lesions as required for the treatment of AF.

This paper is organized as follows. Section II discusses the design of the ETW antenna and the derivation of the GPML ABC for the RS-FDTD technique. Section III presents results obtained from the RS-FDTD-GPML technique and the in-vitro experiment on live bovine hearts to establish the suitability of the proposed ETW antenna for microwave cardiac catheter ablation for the treatment of AF. Finally, important conclusions based on the findings are summarized in Section IV.

II. MATERIAL AND METHOD

A. Derivation of the GPML for TM0n RS-FDTD

The antenna has a rotationally symmetric geometry that is excited by a rotationally symmetric source and, therefore, the resulting electromagnetic field is independent of the cylindrical coordinate \( \phi \). Subsequently, the Maxwell's equations can be expressed as a combination of TE modes which contains \( E_r \) and \( E_z \) components and the TM modes which contain \( H_\phi \) and \( H_z \) components [16]. Since the excitation for the ETW antenna is a TEM mode containing the \( E_r \) and \( E_z \) components, only TM modes will be of interest here. Here, for the sake of completeness, we derive the GPML for the RS-FDTD based on TM0n modes.

By introducing complex coordinate stretching variables \( s_r \) and \( s_z \) in the \( r \) and \( z \) directions, defined in the same way as described in [14], we obtain a split-field GPML formulation in the frequency domain

\[
\begin{align*}
\hat{j} \omega \mu H_{\phi r} &= \frac{\partial E_z}{s_r(\omega) \partial \phi} - \left( \sigma_0^* + \sigma_z^* + \frac{\sigma_0 \sigma_z}{j \omega \mu} \right) H_{\phi r} \quad (1a) \\
\hat{j} \omega \mu H_{\phi z} &= -\frac{\partial E_r}{s_z(\omega) \partial \phi} - \left( \sigma_0^* + \sigma_z^* + \frac{\sigma_0 \sigma_z}{j \omega \mu} \right) H_{\phi z} \quad (1b)
\end{align*}
\]

To transfer the TM0n GPML equations to the time domain, auxiliary variables are introduced for the field terms associated with the inverted \( \hat{j} \omega \) as

\[
\begin{align*}
H_{\phi r}^I &= \frac{1}{\hat{j} \omega} H_{\phi r} \quad (2a) \\
H_{\phi z}^I &= \frac{1}{\hat{j} \omega} H_{\phi z} \\
E_r^I &= \frac{1}{\hat{j} \omega} E_r \\
E_z^I &= \frac{1}{\hat{j} \omega} E_z \quad (2d)
\end{align*}
\]

Equations (2a)–(2d) are obtained by time integration of the corresponding field equations. Substituting (2a)–(2d) into (1a)–(1d) and transferring the resulting equations to the time domain, we obtain

\[
\begin{align*}
\frac{\mu}{\varepsilon} \frac{\partial H_{\phi r}}{\partial t} &= \frac{\partial E_z}{s_r(\omega) \partial \phi} - \left( \sigma_0^* + \sigma_z^* \right) H_{\phi r} - \frac{\sigma_0 \sigma_z}{\mu} H_{\phi r}^I \quad (3a) \\
\frac{\mu}{\varepsilon} \frac{\partial H_{\phi z}}{\partial t} &= \frac{\partial E_r}{s_z(\omega) \partial \phi} - \left( \sigma_0^* + \sigma_z^* \right) H_{\phi z} - \frac{\sigma_0 \sigma_z}{\mu} H_{\phi z}^I \quad (3b) \\
\frac{\varepsilon}{\mu} \frac{\partial E_r}{\partial t} &= \frac{\partial (H_{\phi r} + H_{\phi z})}{s_r(\omega) \partial \phi} - \left( \sigma_0^* + \sigma_z^* \right) E_r - \frac{\sigma_0 \sigma_z}{\varepsilon} E_r^I \quad (3c) \\
\frac{\varepsilon}{\mu} \frac{\partial E_z}{\partial t} &= \frac{1}{r} \frac{\partial (H_{\phi r} + H_{\phi z})}{s_z(\omega) \partial \phi} - \left( \sigma_0^* + \sigma_z^* \right) E_z - \frac{\sigma_0 \sigma_z}{\varepsilon} E_z^I \quad (3d)
\end{align*}
\]

and the auxiliary variables of (2a)–(2d) are expressed in the time domain as

\[
\begin{align*}
\frac{\partial H_{\phi r}^I}{\partial t} &= H_{\phi r} \\
\frac{\partial H_{\phi z}^I}{\partial t} &= H_{\phi z} \\
\frac{\partial E_r^I}{\partial t} &= E_r \\
\frac{\partial E_z^I}{\partial t} &= E_z \quad (4d)
\end{align*}
\]

Equations (3a) and (3c) may now be discretized to obtain the explicit update equations as shown in (5)

\[
H_{\phi r}^{n+\frac{1}{2}}(i,j) = \frac{1 - \left( \sigma_0^* + \sigma_z^* \right) \Delta t}{2 \mu} H_{\phi r}^{n-\frac{1}{2}}(i,j) + \frac{\left( \sigma_0 \sigma_z \Delta t \right)}{2 \mu} \left[ E_z^n(i+1,j) - E_z^n(i,j) \right] - \frac{\sigma_0 \sigma_z \Delta t}{2 \mu} H_{\phi z}^{n}(i,j) \quad (5a)
\]
Equations (3b) and (3d) can also be expressed in the explicit form in a similar way as shown in (5a) and (5b) and, therefore, not repeated here for the sake of brevity.

The auxiliary variables are found by discretizing (4a) and (4c) as follows:

\[ E_{r}^{n+1}(i, j) = \frac{1 - (\sigma_{0}+\sigma_{m})\Delta t}{1 + (\sigma_{0}+\sigma_{m})\Delta t} E_{r}^{n}(i, j) - \frac{\sigma_{0}\Delta t}{2\varepsilon} \left[ H_{0r}^{n+\frac{1}{2}}(i, j) - H_{0r}^{n+\frac{1}{2}}(i, j-1) \right] - \frac{\Delta t}{1 + (\sigma_{0}+\sigma_{m})\Delta t} E_{r}^{n+\frac{1}{2}}(i, j). \]  

(6b)

Equations (3b) and (3d) can also be expressed in the explicit form in a similar way as shown in (5a) and (5b) and, therefore, not repeated here for the sake of brevity.

The auxiliary variables are found by discretizing (4a) and (4c) as follows:

\[ H_{0r}^{n+1}(i, j) = H_{0r}^{n+1}(i, j) + \Delta t \cdot H_{0r}^{n+\frac{1}{2}}(i, j) \]  

(6a)

\[ E_{r}^{n+\frac{1}{2}}(i, j) = E_{r}^{n+\frac{1}{2}}(i, j) + \Delta t \cdot E_{r}^{n}(i, j) \]  

(6b)

and (4b) and (4d) are discretized in a similar fashion.

The parameters \( s_{00}(z) \) and \( \sigma_{z}(z) \) are defined as follows:

\[ s_{00}(z) = 1 + S_{m} \left( \frac{z}{N\Delta z} \right)^{2} \]  

(7)

\[ \sigma_{z}(z) = \sigma_{m} \left( \frac{z}{N\Delta z} \right)^{2} \]  

(8)

where \( N \) is the number of cells used in the PML. The theoretical reflection coefficient may be derived as

\[ R_{th} = \exp \left[ -\frac{2}{\varepsilon c} \int_{0}^{N\Delta z} s_{00}(z)\sigma_{z}(z)dz \right] \]

\[ = \exp \left[ -\frac{2\sigma_{m}N\Delta z(5+3s_{m})}{15\varepsilon c} \right] \]  

(9)

which can be used to choose the value of \( \sigma_{m} \) for a given value of \( s_{m} \) and \( R_{th} \)

\[ \sigma_{m} = -\ln(R_{th}) \frac{15\varepsilon c}{2N\Delta z(5+3s_{m})}. \]  

(10)

B. Validation of RS-FDTD and GPML Technique

To validate the technique, a RS-FDTD grid terminated with a ten-layer PML is compared with a reference solution. The RS-FDTD grid is \( 128\Delta x \times 128\Delta z \), with \( \Delta r = \Delta z = 2.5 \text{ mm} \), \( \Delta t = 4.5 \text{ ps} \) and a total of 500 time steps were computed. The thickness of the PML was ten cells. Equations (7) and
TABLE I

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Antenna A1 and A2</th>
<th>ETW</th>
</tr>
</thead>
<tbody>
<tr>
<td>( \Delta r )</td>
<td>0.08mm</td>
<td>0.015mm</td>
</tr>
<tr>
<td>( \Delta z )</td>
<td>0.25mm</td>
<td>0.25mm</td>
</tr>
<tr>
<td>IMAX</td>
<td>360</td>
<td>460</td>
</tr>
<tr>
<td>JMAX</td>
<td>320</td>
<td>320</td>
</tr>
<tr>
<td>No. of time steps</td>
<td>20,000</td>
<td>20,000</td>
</tr>
<tr>
<td>No. of GPML</td>
<td>10</td>
<td>10</td>
</tr>
<tr>
<td>s(^m)</td>
<td>4</td>
<td>4</td>
</tr>
<tr>
<td>R(^{th})</td>
<td>1x10(^{-4})</td>
<td>1x10(^{-4})</td>
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<tr>
<td>Excitation</td>
<td>Modulated</td>
<td>Modulated</td>
</tr>
<tr>
<td></td>
<td>Sinusoid</td>
<td>Sinusoid</td>
</tr>
</tbody>
</table>

(8) were implemented for the GPML conductivity and coordinate stretching parameters, and the theoretical reflection coefficient used was \(10^{-4}\), with \(s_{m}=5\) and \(\sigma_{m}=(0.367)\). The excitation consisted of an electric current point source located at \((\eta_{r},\eta_{\phi})=\left(1,64\right)\) in the grid, with a Gaussian derivative source function with \(t_{w}=20\) and \(t_{0}=100\)

\[
J_{r}(t) = \frac{2}{t_{w}}(t-t_{0}) \exp\left(-\frac{(t-t_{0})^{2}}{t_{w}^{2}}\right)
\]

The field component is sampled at grid location \((\eta_{r},\eta_{\phi})=\left(37,91\right)\). The reference solution is computed using a large domain in which no reflections from the outer terminating walls appear at the observation point during the 500-step observation period. Fig. 1(a) shows the comparison between the normalized \(E_{r}\) fields of the grid terminated using the GPML with the reference solution. The inset in Fig. 1(a) shows the reflections obtained without GPML to give an indication of the attenuation achieved by the insertion of GPML.

The RS-FDTD-GPML technique is further validated by modeling two existing configurations of microwave catheter antennas reported in [4]. First, two configurations based on the MTM antenna will be analyzed, which will henceforth be denoted as configurations A1 and A2, where according to [4], the dimensions of antenna A1 has \(l = 6.5\) mm, \(t = 4\) mm; and antenna A2 has \(l = 13\) mm, \(t = 4\) mm. The monopoles are immersed in 0.9% saline solution which has a dielectric constant of \(\varepsilon_{r} = 75\) and conductivity of \(\sigma = 2.81\) S/m at 2.45 GHz. The parameters used to model these antennas using RS-FDTD and GPML are given in Table I. The S-parameters for the antennas are extracted from time-domain data using the unmatched port modal extraction technique [14]. Fig. 1(b) shows that the computed S-parameters using RS-FDTD-GPML for antenna A1 and antenna A2 compare well with published measured results. The SAR has also been computed using RS-FDTD-GPML coupled with the discrete Fourier transform (DFT) using the following formula:

\[
SAR = \frac{\sigma}{\rho} \left( |E_{r}|^{2} + |E_{z}|^{2} \right)
\]

where \(\sigma\) is the conductivity, and \(\rho\) is the density of the tissue or phantom material surrounding the antenna. The normalized SAR computed at a distance of \(r = 1.5\) mm from the antenna is plotted for both antenna A1 and A2 in Fig. 1(c) and (d), respectively. These results compare quite well with the results reported in [4] which were obtained using the finite element method (FEM). These results further validate the accuracy of the RSFDTD-GPML technique for the current application.

C. Expanded Tip Wire (ETW) Antenna Design

The configuration of the ETW antenna is shown in Fig. 2(a). The antenna is constructed by first modifying the shield of the coaxial cable and connecting an appropriate metallic tip to the inner conductor so that the antenna becomes an integral part of the cable. A coaxial choke is also placed near the antenna/cable junction. The antenna is fabricated using the Times Microwave System’s TFlex 402 50-\(\Omega\) flexible coaxial cable and is enclosed by a Teflon heat shrink tube fitting through a standard 6F catheter. Under normal circumstances, a standard straight
wire catheter antenna will produce electric-fields (E-fields) perpendicular to the axis of the antenna. The main problem with a standard wire catheter antenna is that it exhibits high level of E-field at the antenna’s tip and antenna/cable junction. The high level of E-field translates to a high level of SAR and, hence, hot spots are created. The higher levels of near E-fields at the distal end of the wire antenna and antenna/cable junction are the result of improper treatment of wire end [17]. When a metallic tip is soldered onto the inner conductor of the wire antenna, the size of the inner conductor at the wire end is effectively increased and, thus, the impedance loading on the wire antenna takes effect [18]. The tip of the wire antenna can be expanded appropriately both in length and width in order to match the input impedance of the wire antenna to that of the coaxial cable connected to the microwave generator. The tip and the choke together act as a balun to help in matching the ETW antenna’s impedance as well as to reduce the flow of leakage currents back into the cable in order to transfer maximum power to the myocardium. Thus, by appropriately choosing the dimensions of choke and tip (C, Coffset, L1, T1, and T2), it is possible to reduce the levels of E-field at the junction of the cable/antenna, and make the E-field confined to the length of the ETW antenna and minimize end radiation. This is shown in Fig. 2(b) where the flow of E-field obtained from the RS-FDTD simulation at steady state is plotted. The arrows indicate the direction of flow of the E-field where as the different grey levels indicate the field strength with darker arrows showing regions of higher E-field intensities. The parameters r1, r2, and r3, as shown in Fig. 2(a), denote the radii of the inner conductor, the dielectric material, and the outer conductor of the antenna, respectively.

Analysis using the RS-FDTD-GPML technique allowed us to investigate in detail the key parameters that affect the antenna performance within myocardium and blood. The antenna parameters have been empirically optimized by modeling the ETW antenna inside a tissue-equivalent phantom. The important design parameters are: 1) L1; 2) C1; and 3) Coffset which affect the antenna near field and resonant frequency in a very complex manner. Fig. 3 shows the effect on the return loss by varying different design parameters of the ETW antenna. Fig. 3(a) shows the effect of the length of the choke (C) on the return loss of the antenna. As C is varied from 5 to 25 mm at 2.45 GHz, it can be seen that the lowest return loss is obtained for C mm. In general, from Fig. 3(a), it can be observed that the ETW antenna provides a reasonable return loss for C in the range of 5 mm and 25 mm. Fig. 3(b) shows that the wire length, L1, is a critical parameter in the impedance matching of the ETW antenna and, in this case we have obtained that L1 = 8 mm provides lowest return loss. Another parameter that is crucial to the impedance matching of the ETW antenna is the offset distance of the choke (Coffset). Again, as shown in Fig. 3(c), the return loss of the ETW antenna varies between 6 and 22 dB as the size of Coffset is varied in the range of 0–16 mm. The L4, T1, and T2 denote the dimensions of the metallic tip of the antenna, where T1 and T2 define the overall radius of the tip while L4 defines the length of the tip. For the proposed ETW antenna, the T1 and T2 are chosen to be slightly smaller than the catheter for easy insertion. The length of the tip L4 is determined by experimental tuning so that a good impedance match is obtained. The final dimensions of the ETW antenna obtained from empirical optimization using the RS-FDTD-GPML technique are...
shown in Table II. A prototype ETW antenna was fabricated based on these values and its return loss was measured with the antenna placed inside a tissue-equivalent phantom. From the measured antenna impedance shown in Fig. 3(d), the antenna input impedance is 55 Ω, which is very close to the impedance of the coaxial cable and the microwave generator (50 Ω), thus achieving a good impedance match at the desired resonant frequency of 2.45 GHz.

Fig. 2(c) shows a comparison between the computed and measured return loss when the prototype ETW antenna is immersed in a tissue-equivalent phantom. Overall, the computed and the measured return loss of the antenna compares very well and, at the resonant frequency of 2.45 GHz, both show lowest possible return loss as desired.

D. Tissue-Equivalent Phantom and In-Vitro Experiment Setup

Tissue-equivalent phantom material comprised of 8.5% TX151 gelling agent, 15% Polyethylene powder, 75% deionized water and 0.5% NaCl was made in order to measure the tissue-equivalent phantom material and the return loss of the ETW antenna were measured using the HP8720 vector network analyzer. For the in-vitro experiments a tissue superfusate bath was constructed using Perspex. The bath was filled with Krebs-Henselite solution bubbled with 95% CO₂ and 5% O₂ to keep the tissue alive. Two tubes, attached to both sides of the bath, connect to a roller pump which regulates the flow of superfusate liquid at a desired flow rate and an external heater keeps the bath, connect to a roller pump which regulates the flow of superfusate liquid at 37 °C. The flow rate of the roller pump was set at 3 liters per minute. Note that the myocardium and the antenna were positioned parallel to the flow of the blood in order to expose the ETW antenna to a larger area of cooling due to the blood flow. This is important as it mimics closely the actual ablation scenario and also helps to test the performance of the antenna under worst case situation. After a lesion is created on the bovine heart tissue, the heart is dissected along the antenna axis so that the lesion length and depth can be obtained directly as shown in Fig. 7. The dissected tissues are immersed in the nitro blue tetrazolium solutions where the dye stains intracellular dehydrogenase that distinguishes viable and necrotic tissues. This is an effective way of identifying the myocardium necrosis and it also helps in measuring the lesion sizes.

III. RESULTS AND DISCUSSION

This section presents computational and measured results for the proposed ETW antenna with a detailed discussion on lesion formation and thermal distribution.

A. Computational Results

The parameters for modeling the ETW antenna are shown in Table II and the RS-FDTD computational domain is given in Fig.2(a). Once the optimal values for the parameters of the ETW antenna were determined, the antenna was constructed and the reflection coefficient measured. A comparison between the measured and computed reflection coefficient is shown in Fig.2(c), which further validates the results obtained from RS-FDTD-GPML technique. Further, the effect of adding a choke to the ETW antenna was also investigated using the RS-FDTD-GPML technique. The choke not only minimizes the leakage currents flowing on the catheter, but also aids in antenna impedance matching to help the antenna deliver microwave energy more efficiently into the myocardium. The results in Fig. 2(c) also indicate that the proposed ETW antenna has a large impedance bandwidth (VSWR<2) which is close to 1 GHz.

For the treatment of AF, arrhythmogenic site of the myocardium has to be heated to a temperature well over 50 °C in order to create tissue necrosis [5]. Hence, during the ablation process, the dielectric constant of the tissue surrounding the antenna changes with the increase in the temperature of the myocardium. The change in the dielectric constant can alter the resonant frequency of the antenna and thereby the impedance matching of the antenna will be affected. The large impedance bandwidth for the ETW antenna can potentially ensure that the performance of the antenna is not greatly affected by the change of dielectric constant due to heating of the myocardium. Such feature is necessary since it is not possible to directly monitor the tissue ablation process during microwave ablation.

The SAR of the ETW antenna computed using the RS-FDTD-GPML technique is shown in Fig. 4. As discussed above, other than impedance matching, the effect of adding a choke to the ETW antenna is to stop the E-field from propagating down the cable. Fig. 4(a) shows the SAR of ETW antenna without the choke and Fig. 4(b) shows the SAR of ETW antenna with a choke. From Fig. 4(a) and (b), it can be clearly seen that the choke significantly reduces the backward propagation of E-field along the catheter as indicated by low SAR values at the cable-antenna junction, and thus, minimizing unnecessary cable heating.

B. In-Vitro Experiment Results

As discussed in Section II-C, the tissue superfusate bath was filled with tissue-equivalent phantom and temperature sensing liquid crystal sheets were used to obtain a visual thermal profile of the ETW antenna. The thermal profile of the ETW antenna, Fig. 4(b), shows that the temperature profile of the ETW antenna is very well confined within the antenna section. Very little temperature rise can be seen on the coaxial cable indicating that the E-field along the catheter surface is very low. An applied power of 60 W for 30 s was used to obtain the temperature distribution shown in Fig. 4(b) on a temperature sensitive liquid crystal sheet.

C. SAR Calculation From Measured Temperature

Once the thermal profile of the ETW antenna was obtained using the liquid crystal sheet, the myocardium tissue phantom was replaced with live bovine heart tissue. The temperature rise
due to antenna radiation was measured in real-time using the Luxtron 3100 fiber-optic temperature sensing system. A total of four fiber optic thermocouples were used to record the temperature rise and these thermocouples were placed at 1, 4, 7, and 10 mm deep into the myocardium perpendicular to the antenna axis. Initially, a microwave power of 20 W at 2.45 GHz was applied to the myocardium tissue for 30 s and the temperature changes were recorded. By calculating the rate of the temperature rise in the linear region of the temperature curve, the SAR value at a particular point within the tissue can be determined using the following equation [8], [10]:

\[
\text{SAR} = \frac{W}{kg} = c_p \frac{\Delta T}{\Delta t}
\]

(13)

where the \(c_p\) is the specific heat of the myocardium tissue and \(\Delta T/\Delta t\) is the rate of temperature change over a short period of time. The measured temperature data usually contains temperature rise due to the energy deposited by the antenna and heat conduction from the surrounding tissue. Hence, to calculate the energy deposited due to antenna alone using (13), one must be cautious in obtaining the temperature data so that the temperature rise due to heat conduction is minimized in data collection. As indicated in [8], the power level chosen should only be sufficient enough so as to cause a temperature change within the myocardium tissue. If the power is exceedingly high, the temperature at a given point within the myocardium tissue may not represent the true temperature rise due to microwave energy deposited by the catheter antenna since the heat conducted from the neighboring tissues may contribute significantly to the temperature rise. On a similar note, the temperature measurement should only be made at the linear region of the temperature curve. The temperature measurement duration should also be as short as possible as to avoid heat conduction from the neighboring myocardium tissues. These two factors will affect the accuracy of calculated SAR when using (13). From in-vitro experiments, we found empirically at 2.45 GHz that, without cooling from circulating blood, a combination of 20 W of microwave power for 20 s gave the most accurate measured temperature for calculating SAR. However, the combination of ablation power

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Fig. 4. The simulated SAR of the (a) ETW without choke and (b) ETW with choke antennas. (c) Thermal distribution of the ETW antenna within tissue-equivalent phantom for 60 W of applied power. The dashed line outlines the thermal distribution of the ETW antenna.
and duration will be dependent on the type of antenna used and possibly the operating frequency of the microwave generator.

Using the measured rate of temperature change at 1 mm below the surface of the myocardial tissue, the SAR of the ETW antenna was calculated using (13), which will henceforth be denoted as the “measured” SAR. These measured SAR values are compared with the computed SAR values from the RS-FDTD-GPML technique and are plotted in Fig. 5.

From Fig. 5, it can be observed that the SAR values along the surface of the catheter are very low indicating that the surface heating on the catheter is very small. Further, the predicted SAR has shifted slightly (less than 5%) toward the antenna-cable junction where as the measured SAR is concentrated on the centre of the ETW antenna. Comparing with Fig. 4(b), a higher concentration of predicted SAR values can be observed at the antenna-choke junction as shown by the computed SAR is very good agreement with the measured values.

D. Determination of Optimum Power Setting for Microwave Ablation

Unlike RF ablation wherein the contact impedance at the tip of the RF electrode increases with the size of the lesion for a fixed time interval, ablation using microwave energy can provide no such measurable control parameter. Therefore, knowledge of the optimum microwave power required to generate a desirable sized lesion needs to be determined prior to catheter ablation. This can be determined by examining the power-temperature profile of the microwave antenna, which may be defined as the rate of temperature rise with the increase in applied power for a predefined time. From this profile, the optimum energy setting can be determined.

The power-temperature profile of the ETW antenna is shown in Fig. 6(a) with ablation duration fixed at 30 s. Examination of the power-temperature profile show that, as expected, there is a sharp increase in temperature as the power is increased from 60 to 100 W. However, the rate of temperature increase at distances 7 and 10 mm below the tissue surface due to 100 W of power is not as high as with 80 W. The reason for this is that at 1 mm and 4 mm below the surface of the antenna, the heating is mainly due to direct electromagnetic energy deposition. However, at a depth of 7 and 10 mm, the amount of direct electromagnetic energy deposited by the antenna is less compared with the energy deposited at the surface of the myocardium. Therefore, tissue heating in areas deep down the myocardium relies mostly on the heat conducted from other areas of myocardium in addition to the residual electromagnetic energy from the antenna. At lower power levels, for example, below 40 W, the rate of temperature increase is almost same at the surface as well as deep into the myocardium as shown in Fig. 6(a). If a microwave power level of 40 W or less is used, the time it takes to create a transmural lesion is longer and during this time the lesion can grow unnecessarily wide destroying the surrounding healthy tissues. However, when a higher power level greater than 80 W is used, the tissue at the surface of the myocardium is heated more quickly than the tissues deep inside the myocardium. This is clear from
Referring to Fig. 7, the lighter tissues are necrotic tissues whereas the darker tissues represented the normal tissues unaffected by the cable heating. The use of applied microwave power levels of 40, 60, 80, and 100 W were used to create lesions in the ventricle of the ETW antenna. Applied microwave power settings to create lesions with ETW antenna without the spiral antenna. From this, it is fair to conclude that, for a given power level and duration, the ETW antenna is capable of depositing more electromagnetic energy into the myocardium than the spiral antenna.

Live bovine heart muscles were used in the in-vitro testing with the ETW antenna. Applied microwave power levels of 40, 60, 80, and 100 W were used to create lesions in the ventricle of the bovine heart. The superior impedance match and low field levels at the cable/antenna junction allowed us to use higher power settings to create lesions with ETW antenna without having to worry about excessive cable heating. The use of shorter ablation duration also helped to keep the cable heating to be minimal. The representative lesions are shown in Fig. 7 and each grid measures a 10 mm x 10 mm cell. Referring to Fig. 7, the lighter tissues are necrotic tissues whereas the darker tissues represented the normal tissues unaffected by the microwave radiation. The measured depths of the lesions are summarized in Table III. Fig. 7 shows that the lesions are continuous and at 80 and 100 W, their depth is greater than 5 mm which is adequate for the ablation in the atrium. Since the radiation from ETW antenna is perpendicular to its axis, long and linear lesions can be achieved with minimum antenna movement or relocation.

### IV. CONCLUSION

In this paper, we proposed a novel ETW antenna to deliver highly localized microwave power to the myocardium for the treatment of AF. It is shown that the ETW antenna is very well matched to the myocardium at the operating frequency of 2.45 GHz. The RS-FDTD-GPML technique has been used to efficiently model the interaction of antenna with myocardium as well as to empirically optimize antenna parameters for best performance.

The proposed ETW antenna is designed for insertion through a 6F catheter into the myocardium and is constructed in such a way that it forms an integral part of the coaxial cable for ease of fabrication and to ensure antenna’s physical integrity during catheter insertion and ablation. We have demonstrated that by choosing appropriate size of the choke and the tip, it is possible to confine the electric-field around the length of the ETW antenna which resulted in localized deposition of microwave energy deep into the myocardium. The results from the in-vitro experimental study also confirm the above findings. Further, the measured lesion sizes establish the effectiveness of the proposed antenna to create linear transmural lesions as required for the treatment of AF.

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