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Transthoracic Atrial Defibrillation Energy Thresholds are Correlated to Uniformity of Current Density Distributions

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Abstract—**Previous studies have shown that successful defibrillation depends on the uniformity of current density in the heart and the percentage of total current reaching the heart. This study uses an anatomically-realistic finite element computer model of the human torso for external atrial defibrillation to (1) examine the defibrillation energy thresholds and current density distributions for common clinical paddle placements and (2) investigate the effects of electrode shifts on these defibrillation parameters. The model predicts atrial defibrillation threshold (ADFT) energy by requiring a voltage gradient of 5 V/cm over at least 95% of atrial myocardium. This study finds that variation in electrode placement by only a few centimeters increases ADFTs by up to 46% with a corresponding change of 38% between the average current density in the left and right atria and 34% between the heterogeneity indices of atrial current density distributions. Additionally, the heterogeneity index, or degree of uniformity, is linearly correlated to the ADFT (R^2=0.9). We suggest that uniformity of current density distribution, in addition to minimum current density, may be an important parameter to use for predicting successful defibrillation when testing new electrode placements.**

I. Introduction

ATRIAL fibrillation (AF) affects two million people in the United States. AF patients are at an increased risk of developing blood clots and embolic strokes. In fact, 15% of all strokes occur in patients with AF [1]. Treatments for AF include pharmacological therapy, electrical cardioversion, radiofrequency ablation, and atrial pacemakers.

Electrical cardioversion is used to revert the atria back to sinus rhythm in patients with both acute and chronic AF. Computer modeling of defibrillation has been validated for both external and internal ventricular defibrillation [2-7]. Camacho et al. developed a three-dimensional finite element model of the human thorax to examine current density distributions during transthoracic defibrillation. The study suggests that the spatial distribution of myocardial current density depends on electrode paddle placement [2]. Karlon et al.developed a realistic finite element model of the canine thorax to study ventricular defibrillation, and this model compares favorably with in vivo experiments [6, 7]. Panescu et al. validated a similar three-dimensional finite element model of the human thorax by comparing predicted defibrillation values to experimental data from a human subject and further concluded that current density distribution depends on electrode placement and size [5]. A three-dimensional finite element model of both transthoracic and transvenous defibrillation constructed by Jorgenson et al. found that a shift in electrode placement of less than 2 cm caused a 12% change in average myocardial voltage gradients [4]. All of these models use the critical mass hypothesis [8] to predict successful defibrillation, i.e. a critical mass of myocardial tissue (ranging from 75-95%) must be above a minimum voltage gradient.

Previous studies have shown that successful defibrillation depends on the uniformity of current density in the heart and the percentage of total current reaching the heart [2, 5-7]. The purposes of this study are to use a three-dimensional, anatomically realistic, finite element, computer model of the human torso for external atrial defibrillation to (1) examine the defibrillation thresholds and current density distributions for common clinical paddle placements and (2) investigate the effects of electrode shifts on these defibrillation parameters. The uniformity of current density in the atria and normalized ADFTs are compared for each electrode placement.

II. Methods

A. Model Construction and Formulations

This study was performed using a volume conductor model of the human torso, previously developed for external atrial defibrillation simulations [9]. The realistic geometry for this study was extracted from 90 transverse MRIs. The images (256x256 pixels) were acquired at 5 mm separation. Image acquisition coincided with end diastole. The following tissue structures were digitized from the MRIs: body surface, fat, skeletal muscle, lungs, great vessels of the heart (aortic arch, ascending and descending aorta, pulmonary trunk, and superior vena cava), epicardium, left atrium, right atrium, left ventricle, and right ventricle [9].

The unstructured mesh contains 60,000 nodes and 381,101 tetrahedra. Each tetrahedral element is defined as a member of its corresponding region at the completion of the meshing process. Triangulated surfaces of the model are shown in Fig. 1.

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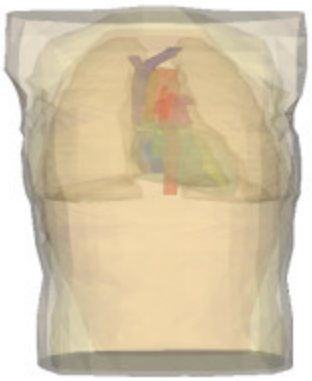

Fig. 1. Anterior view of triangulated surfaces of the model. The body surface, the fat, the lungs, the heart, and the great vessels are shown.

Each volume element is assigned a conductivity tensor, according to the tissue type of the element's respective region. All regions are treated as isotropic for the entire study. The heart chambers and vessels are assigned the conductivity of blood. Tissue conductivity values are shown in Table 1[10-13].

TABLE I
TISSUE CONDUCTIVITIES

Tissue Type	Conductivity (mS/cm)
Lungs	0.78 [13]
Blood	6.67 [10]
Myocardium	2.50 [10]
Connective Tissue	2.22 [10]
Fat	0.50 [12]
Skeletal Muscle	2.50 [11]

With knowledge of the applied voltage, defibrillation electrode location, and tissue conductivity, the model computes potential gradient distributions using the finite element method. The potential in a volume conductor can be represented by:

$$\nabla \cdot \sigma \nabla \Phi = 0 \quad (1)$$

where σ represents the conductivity tensor and Φ is the electric potential. Nuemann boundary conditions apply on the torso surface. Equation (1) is solved by the finite element method to determine the potential at each node, applied current at the electrode nodes, and voltage gradients in atrial volume elements. ADFT is calculated from the energy relation $W_e = \frac{1}{2} CV^2$. V represents the potential difference required to produce a minimum potential gradient (5 V/cm) throughout a critical mass (95%) of the atrial myocardium. The capacitance of the pulse generator, C, is assumed to be 140μF for this study. The current density is obtained by multiplying the element conductivity value with the electric field (potential gradient).

B. Investigation of Defibrillation Parameters

The clinical model for this study consists of rectangular electrodes (90 cm^2) in clinical anterior-anterior (AA1) and anterior-posterior (AP1) placements [3] (Figure 2). The effect of electrode location is determined by shifting AA1 and AP1 2-3 cm to produce six additional electrode placements (AA2-AA4, AP2-AP4). For AA positions, the apical electrode is held stationary and the basal electrode is shifted. For AP positions, the posterior electrode is stationary and the anterior electrode is shifted (Figure 2).

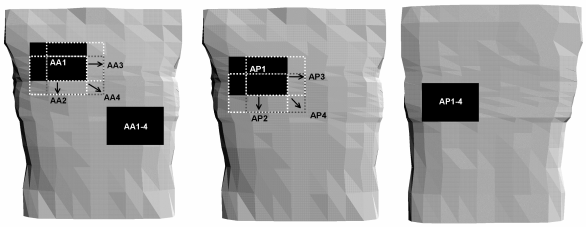

Fig. 2. Anterior electrodes for the baseline anterior-anterior (AA1) and anterior-posterior (AP1) in black and shifts (AA2-AA4, AP2-AP4) indicated by dotted lines.

For each simulation, current densities are calculated for the volume elements of both the right and left atria. In order to examine the uniformity of current density in the atria, histograms of the current density in both the left and right atria were examined. From these histograms, the median (P_{50}), inhomogeniety ($P_{5\text{-}95}$), and the heterogeneity index ($P_{5\text{-}95}/P_{50}$) were calculated [14]. In addition, the percent difference of the volume-weighted average current density between the left and right atria was calculated for each placement.

III. RESULTS

The clinical electrode placements selected correspond to the two placements used in a clinical study [15]. These baseline AA1 and AP1 positions yield ADFTs of 18.9 J and 15 J, respectively. All ADFTs for this study are normalized to the ADFT for AA1. The results compare favorably with [15], which found that the AP placement required 20% less energy to successfully defibrillate. AP1 also has a more uniform overall current density than AA1 (Table 2). Differences in ADFT vary from 14% to 46% for AA1 and 0% to 15% for AP1.

TABLE 2
CURRENT DENSITY COMPARISONS

Electrode Placement	ADFT	% Difference RA-LA	$P_{5\text{-}95}/P_{50}$
AA1	1.00	24.4	0.623
AA2	1.42	37.1	0.843
AA3	0.86	24.5	0.622
AA4	1.46	38.5	0.939
AP1	0.79	16.2	0.528
AP2	0.67	11.8	0.517
AP3	0.79	4.8	0.425
AP4	0.84	22.5	0.589

A. *Current Density Distribution*

Table 2 shows the percent difference of the volume-weighted average current density between the left and right atria and the heterogeneity index for the atria. For all placements except AP3, the percent difference between the left and right atria of the weighted average current density is higher for placements with higher ADFTs. Current density heterogeneity indices in the atria are also higher for placements with higher ADFTs, except for AP3. Linear regression was performed to determine the relationship between ADFT and each of the measures of uniformity of current density. The R^2 values were 0.90 for heterogeneity index and 0.80 for the percent difference between the right and left atria. Regression lines for the heterogeneity index and the percent difference are shown in Figures 3 and 4. Figures 5 and 6 show current densities for a cross section of the right atria for AA1 and AA4, respectively.

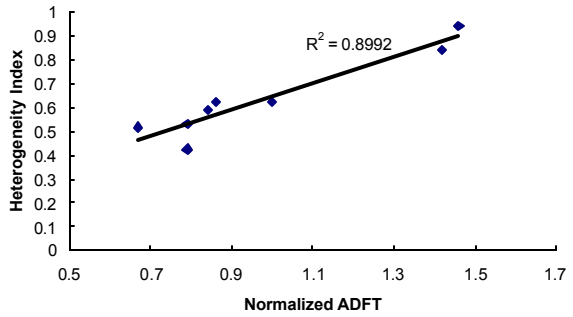

Fig. 3. Heterogeneity indeces of the current density in the atria versus normalized ADFTs for the eight electrode placements.

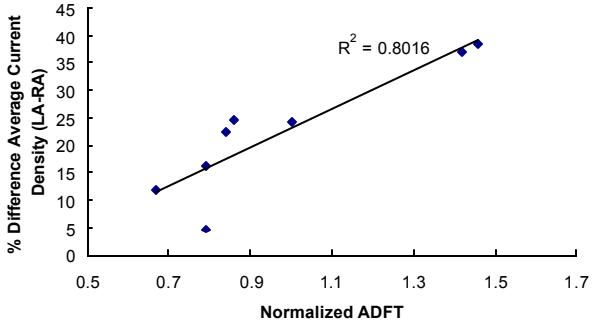

Fig. 4. Percent differences of the average current density in the left and right atria versus normalized ADFTs for the eight electrode

IV. Discussion

The clinical placements AA1 and AP1 represent the two placements investigated in a clinical study by Botto, which found that the AP placement required 20% less energy than AA to successfully defibrillate [15]. The clinical model of this study finds that AP1 requires approximately 20% less energy than AA1 to achieve defibrillation. This suggests that the model can predict relative differences in ADFTs among different electrode placements. AP1 also has a more

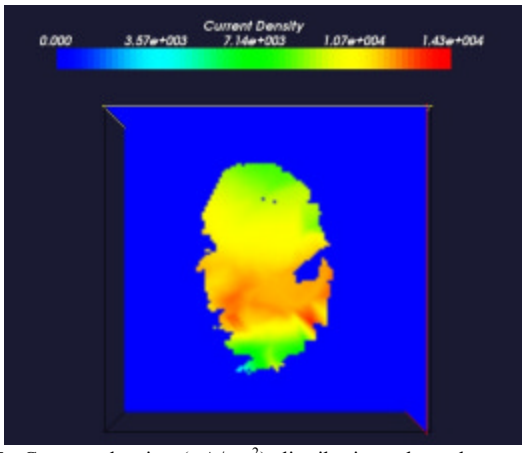

Fig. 5. Current density ($\mu A/cm^2$) distributions throughout a cross section of the right atrium for AA1. The percent difference between the average current density in the right and left atria is 24.4%. The heterogeneity index is 0.623.

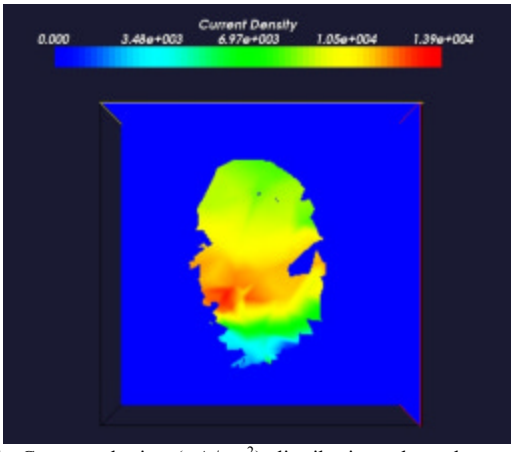

Fig. 6. Current density ($\mu A/cm^2$) distributions throughout a cross section of the right atrium for AA4. The percent difference between the average current density in the right and left atria is 38.5%. The heterogeneity index is 0.939.

uniform overall current density than AA1 (Table 2). This agrees with previous research indicating that a more uniform current density distribution corresponds to a lower DFT [2, 5-7].

A. *Current Density Distribution*

Our heterogeneity index of current density uniformity shows a strong relationship with ADFT (R^2 = 0.90). However, the relationship between the percent difference of volume-weighted average current density between the left and right atria is not as strong (R^2 = 0.80). This lower R^2 is due the value for AP3. The AP3 placement has a lower average current density than the other AP placements, but also a lower heterogeneity index. The ADFT for AP3 falls in the middle of the range for AP placements. Nonetheless, our measures of heterogeneity support the idea that successful defibrillation is correlated to the uniformity of current density in the heart.

B. Limitations

Our model is limited by the accuracy of the selected tissue conductivities and isotropic tissue properties. An anisotropic representation of the heart tissue would change the current density distributions and could affect our heterogeneity indices. Our assumption of critical mass is supported experimentally for ventricular defibrillation [16], but there is no empirical data to support this assumption for the atria. ADFTs calculated in this study are not clinically accurate, even though the relative differences in ADFT agree with clinical data. There may be a parameter missing from the external atrial defibrillation model, such as tissue-electrode interface impedance, that causes the ADFT magnitudes to be much lower than clinically reported values.

This model cannot account for some additional factors that may affect ADFTs such as the type of shock waveform applied (monophasic or biphasic) patient's history of antiarrhythmic drug treatment [17], and duration of AF [15, 18]. In addition, there is a lack of clinical studies reporting atrial defibrillation thresholds for direct comparison. Recent studies comparing monophasic and biphasic shock waveforms have shown that biphasic shocks significantly improve atrial defibrillation success rates up to 99% [19, 20]. However, this high level of success still requires multiple shocks since defibrillation attempts are commonly delivered with a step-up protocol.

Electrode placement optimization could prevent some failed defibrillation shocks and reduce skin-related morbidity, especially for those patients with longer duration AF.

V. Conclusions

In our investigation of the uniformity of current density distribution during atrial defibrillation shocks, we find that atrial defibrillation thresholds depend on both the uniformity of current density in the atria and the differences in average current density magnitude between the atria. We suggest that the uniformity of current density distributions in the atria, in addition to minimum current density, may be an important parameter to use for predicting successful defibrillation when testing new electrode placements.

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