Defibrillation Efficacy of Different Electrode Placements in a Human Thorax Model

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Running title: Modeling Transvenous Defibrillation

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Abstract

The objective of this study is to determine the defibrillation threshold (DFT) for different electrode placements using a three dimensional physiologically realistic finite element model of the human thorax. Coil electrodes (Endotak DSP, model 125, Guidant/CPI) are placed in the RV apex along the lateral wall (RV), pulled back proximally 10 mm from the RV apex along the lateral wall (RVprox), in the RV apex along the anterior septum (RVseptal), and in the SVC. An active pulse generator (Can) is placed in the subcutaneous prepectoral space. Five electrode configurations are implemented: RV→SVC, RVprox→SVC, RVseptal→SVC, RV→Can, and RV→SVC+Can. DFTs are defined as the energy required to produce a potential gradient of at least 5 V/cm in 95% of the ventricular myocardium. DFTs for RV→SVC, RVprox→SVC, RVseptal→SVC, RV→Can, and RV→SVC+Can are 10, 16, 7, 9, and 6 J, respectively. The calculated DFTs for each configuration fall within one standard deviation of the mean DFTs reported in clinical studies using the Endotak leads. The relative changes in DFT between electrode configurations also compare favorably. The advantage of the computer model is that any number and type of electrode placement can be analyzed to determine the DFT or other defibrillation parameters, saving both time and cost of clinically evaluating different electrode configurations.

Keywords: ventricular defibrillation, implantable cardioverter defibrillator, lead systems, defibrillation modeling, defibrillation threshold
Introduction

The implantable cardioverter defibrillator (ICD) has proven to be highly effective in protecting against sudden cardiac death.\textsuperscript{1,2} In the first year of treatment, patients with ICDs experience a 38\% reduction in mortality compared to those taking anti-arrhythmic drugs.\textsuperscript{2} The improvement in defibrillation therapy over the past 10 years is due not only to clinical studies but also to the application of modeling techniques. Computer modeling of defibrillation has been performed for and validated with experimental measurements in external and internal defibrillation.\textsuperscript{3-7} The achievements of these previous modeling studies have proven the utility of realistic models to accurately predict electric fields during defibrillation by validating not only computed defibrillation threshold (DFT) voltages, currents, and impedances,\textsuperscript{3-6} but also computed electric fields at known measurement locations throughout the heart and thorax.\textsuperscript{7}

To obtain information about the spatial distribution of the shock field, investigators have used sock electrodes\textsuperscript{8,9} to map the epicardium as well as plunge needle electrodes\textsuperscript{8} to measure the shock field transmurally. Studies have shown that immediately following defibrillation shocks, electrical activity first appears in an area of the heart subjected to the weakest portion of the electric field.\textsuperscript{8} The disadvantage of these mapping techniques is that they do not provide detailed descriptions of the electric fields throughout the entire myocardium. Computer models have the capability to provide detailed, three-dimensional information about defibrillation fields in the myocardium without perturbation of the field by the measurement system.

The primary purpose of this study is to validate our human thorax model for specific electrode placements using the Endotak DSP (model 125) lead system manufactured by Guidant/CPI. A secondary purpose of this study is to display the utility of the computer model. The computer model is useful not only for investigating the influence of parameters such as
electrode placement and size on DFTs, but also for exploring mechanisms of defibrillation by examining the electric fields produced during a shock.

**Methods**

**Model Construction**

The volume conductor model for this study is constructed from 90 transverse magnetic resonance images (MRI) of a human torso. The healthy, human male subject was approximately 190 cm tall and weighed approximately 80 kg. The images of the torso were acquired at 5 mm separation using T1 weighted imaging on a 1.5 Tesla Siemens Magnetom system. Image acquisition was gated to coincide with end diastole to simulate ventricular volume during ventricular defibrillation. The software program *Image 1.60* distributed by the National Institutes of Health is used to display the images and manually digitize the following tissue surfaces of interest for each image: body surface, skeletal muscle, lungs, epicardium, left atrium, right atrium, left ventricular endocardium, right ventricular (RV) endocardium, aorta, superior vena cava (SVC), and pulmonary trunk. Cubic spline algorithms\(^{10}\) are used to generate a set of equidistantly spaced points for each tissue surface (i.e. epicardium). From a knowledge of the three-dimensional coordinates selected to represent each surface, a triangulation algorithm was developed to connect the points on each respective tissue surface to generate a set of closed surfaces.

Volume points are subsequently generated and connected in the form of tetrahedral elements.\(^{11,12}\) The mesh generation process is completed by defining each tetrahedron as a member of its respective tissue type (i.e. myocardium) and assigning the corresponding conductivity. The torso model for this study is comprised of 33,992 nodes and 214,957
tetrahedral elements (Figure 1). The conductivities of the interior thorax used in the model are shown in Table I. For purposes of this study, all tissues are considered to be isotropic, where the conductivity of tissue fibers is the same in all directions. The conductivity of blood is assigned to each of the four chambers and great vessels of the heart.

Defibrillation Electrode Placement

Catheter electrodes are modeled as cylindrical isoelectric volumes. The distal and proximal electrodes are 2.73 mm in diameter and 48 and 68 mm in length, respectively. The results are leads identical to the Endotak DSP (model 125) leads manufactured by Guidant/CPI. The distal electrode is placed in the RV apex along the lateral wall (RV), pulled back proximally 10 mm from the RV apex along the lateral wall (RV$_{\text{prox}}$), or in the RV apex along the anterior septum (RV$_{\text{septal}}$). The proximal electrode is positioned in the SVC so that the proximal end of the electrode is situated at the brachiocephalic branch of the SVC. The ICD is implanted subcutaneously in an infracavicular position. The active can is modeled as a rectangular isoelectric volume and is 58x58x16 mm. Five electrode configurations are implemented: RV→SVC, RV$_{\text{prox}}$→SVC, RV$_{\text{septal}}$→SVC, RV→Can, and RV→SVC+Can. Figure 2 shows the electrode placements in the torso. The electrodes are incorporated into the model by first constructing the surface of each type of electrode and then including them with the tissue surfaces on the front end of the mesh generation process.

Formulations for the Model

The finite element method$^{16}$ is used to compute the electric field intensity within the heart from a knowledge of defibrillation shock strength, defibrillation electrode location, and the
relative conductivities of the interior thorax. It has been reported that over 90% of the ventricular mass should be exposed to a minimum potential gradient of 5 V/cm for successful defibrillation.\textsuperscript{8} Based on these findings, the DFT for this study is defined as the energy required to produce a potential gradient of at least 5 V/cm throughout 95% of the ventricular myocardium. Thus, weak defibrillation fields are defined as potential gradients less than 5 V/cm.

In a previous modeling study, Jorgenson et al.\textsuperscript{7} determined that the accuracy of the finite element solution can be improved by incorporating the impedances of the lead wires and electrode-tissue interfaces into the model. A value of 10 Ω is assumed for the combined interface and lead impedance at the anode and cathode.\textsuperscript{7,17} The DFT for this study is calculated using the energy relation $W_e = \frac{1}{2} CV^2$. $C$ is the capacitance of the pulse generator used in combination with the Endotak leads ($C = 140$ μF). $V$ is the potential difference between electrodes required to produce a minimum potential gradient of 5 V/cm throughout a critical mass of 95% of the ventricular myocardium.

**Results**

DFTs for RV→SVC, RV\textsubscript{prox}→SVC, RV\textsubscript{septal}→SVC, RV→Can, and RV→SVC+Can are 10, 16, 7, 9, and 6 J, respectively. Clinical studies using the Endotak leads modeled in this investigation have reported DFTs (mean±SD) of 11±3, 10±4, 10±5, and 8±4 J for RV→SVC, RV\textsubscript{septal}→SVC, RV→Can, and RV→SVC+Can, respectively.\textsuperscript{18,19} The calculated DFTs for each configuration fall within one standard deviation of the mean DFTs reported in these clinical studies. Direct comparison between computed and clinical DFTs cannot be made for RV\textsubscript{prox}→SVC because the only data available for this configuration has been obtained from pigs.\textsuperscript{20} Nevertheless, our results indicate that the DFT is significantly higher for RV\textsubscript{prox}→SVC
compared to RV→SVC which is consistent with the experimental findings of the aforementioned study. Table 2 summarizes the clinical and computed DFTs for each electrode configuration.

The potential and potential gradient fields on the RV and LV endocardium for RV→SVC+Can are shown in Figure 3. The maximum potential and potential gradient on these surfaces occur nearest to the shocking electrode in the RV. Figure 4 displays the weak fields (potential gradients < 5 V/cm) in the myocardium for RV\textsubscript{prox}→SVC and RV→SVC+Can, which produce the highest and lowest DFTs, respectively, for the electrode placements analyzed in this study. Visual comparison of the weak fields for these two configurations shows that the weak field is more concentrated, or contiguous, for RV\textsubscript{prox}→SVC than the weak field for RV→SVC+Can.

**Discussion**

The comparative differences in DFTs between electrode configurations predicted with the torso model are consistent with reported clinical studies. In addition, the calculated DFTs for each configuration fall within one standard deviation of the mean DFTs reported in clinical studies using the Endotak leads. Weak shock field distributions in the myocardium were shown for two electrode configurations to display the ability of the model to present useful information not readily obtained by experimental studies. Weak fields were compared between the least efficacious lead system (RV\textsubscript{prox}→SVC) and most efficacious lead system (RV→SVC+Can) in this study. For RV\textsubscript{prox}→SVC, the weak field is spatially contiguous, i.e. concentrated at the apex and extending throughout the LV lateral free wall. The weak field for RV→SVC+Can is spatially irregular, i.e. spread out over the posterior myocardium with many perforations. The results indicate that the spatial distribution of the weak field may be directly related to the
efficacy of the lead system. Further examination of weak fields for other electrode configurations is needed to determine if the spatial distribution of the weak field could be used as a measure of defibrillation efficacy when developing and testing new strategies to lower DFTs. Determining the degree of spatial irregularity in the weak fields requires visual, qualitative inspection and is therefore a subjective process. Future plans include quantifying the spatial irregularity or continuity of the weak fields.

The significance of the study is that computer models allow for a direct way to investigate the influence of defibrillation parameters such as electrode placement on defibrillation efficacy, saving both time and cost of clinically evaluating different electrode configurations. In addition, the computer model provides the capability to examine in three-dimensional detail the distribution of defibrillation fields, such as weak potential gradient fields. Given that our model has been validated with clinical data, we believe it can give valuable insight into the mechanisms of defibrillation by correlating defibrillation efficacy to the distribution of defibrillation fields in the heart.
References
2. The Antiarrhythmics Versus Implantable Defibrillators (AVID) Trial Executive Committee: Are implantable cardioverter-defibrillators or drugs more effective in prolonging life? Am J Cardiol 1997; 79: 661-663.
### Table I.

Tissue Conductivities

<table>
<thead>
<tr>
<th>Tissue Type</th>
<th>Conductivity (mS/cm)</th>
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<tr>
<td>Lungs$^{13}$</td>
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<tr>
<td>Blood$^{14}$</td>
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<tr>
<td>Myocardium$^{14}$</td>
<td>2.50</td>
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<tr>
<td>Connective Tissue$^{14}$</td>
<td>2.22</td>
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<tr>
<td>Skeletal Muscle$^{15}$</td>
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### Table II.

Comparison of Clinical and Computed DFTs

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<tr>
<th>Configuration</th>
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<th>Computed DFT (J)</th>
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<tbody>
<tr>
<td></td>
<td>Reference</td>
<td>DFT ± SD (J)</td>
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<tr>
<td>RV→SVC</td>
<td>Gold et al.$^{18}$</td>
<td>11±3</td>
</tr>
<tr>
<td>RV$_{prox}$→SVC</td>
<td>**</td>
<td>**</td>
</tr>
<tr>
<td>RV$_{septal}$→SVC</td>
<td>Winter et al.$^{19}$</td>
<td>10±4</td>
</tr>
<tr>
<td>RV→Can</td>
<td>Gold et al.$^{18}$</td>
<td>10±5</td>
</tr>
<tr>
<td>RV→SVC+Can</td>
<td>Gold et al.$^{18}$</td>
<td>8±4</td>
</tr>
</tbody>
</table>

**No human data available using Endotak leads for RV$_{prox}$→SVC.
Figure 1. Anterior views of the torso model. (a) All tissue surfaces included in the model are shown. (b) The epicardium, RV and LV endocardium, right atrium (RA), left atrium (LA), SVC, aorta and pulmonary trunk are shown.
Figure 2. Anterior views of electrode placements. (a) The RV, SVC, and Can electrodes are shown. The skeletal muscle and lungs are not shown. (b) The RV chamber and the three different placements of the RV electrode are shown.
Figure 3. (a) Potential fields and (b) potential gradient fields resulting from a defibrillation strength shock on the RV and LV endocardial surfaces. Left-lateral views of the endocardial surfaces are shown. The RV is on the left and the LV is on the right side of (a) and (b). The legends for the potential field and potential gradient field in (a) and (b) are given in volts (V) and V/cm, respectively. The DFT for this configuration is 6 J with an applied DFT potential difference of 297 V.
Figure 4. Weak fields for $RV_{prox} \rightarrow SVC$ and $RV \rightarrow SVC+Can$. The left-hand column are anterior views and the right-hand column are posterior views. The epicardium, left and right atria, and RV and LV endocardium are shown as transparent surfaces. The RV electrode position is also shown. The dark regions represent the areas of weak fields, which are defined as those regions with potential gradients $< 5 \text{ V/cm}$. For $RV_{prox} \rightarrow SVC$, the locations of the weak fields are apical and left lateral. For $RV \rightarrow SVC+Can$, the weak fields are apical and posterior. The weak field is near the epicardium for $RV \rightarrow SVC+Can$ but extends transmurally from the apical LV endocardium to the epicardium for $RV_{prox} \rightarrow SVC$. 