Three-dimensional Potential Gradient Fields Generated by Intracardiac Catheter and Cutaneous Patch Electrodes

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Background. Defibrillation may be improved if electrode configurations can be found that create a larger and more even voltage gradient field across the heart. This study determined the magnitude of the shock gradient fields generated by four nonthoracotomy electrode configurations for defibrillation.

Methods and Results. In six dogs, a catheter was inserted containing a right ventricular apical electrode (V) and a right atrial electrode (A). A cutaneous patch electrode (P) was placed on the left lateral thorax. Shock potentials were recorded simultaneously from 128 electrodes in the left ventricular and right ventricular subepicardium and subendocardium, ventricular septum, and atria. With the chest closed, 50-mA shocks were given during diastole via the following lead configurations: V→A (V, cathode; A, anode), V→P; V→A+P; and V+A→P. Potential gradients were calculated at the subepicardium and subendocardium in millivolts per centimeter per volt of shock. In most dogs, the V→A+P configuration produced higher gradients throughout the ventricles than did V→A, V→P, or V+A→P. The maximum potential gradient was smaller for the V+A→P configuration than for V→A, V→P, or V→A+P. The gradient fields for the configurations with the catheter alone or combined with P were uneven.

Conclusions. It is possible to estimate shock gradient fields in three dimensions. Of the four configurations tested, V→A+P produced the highest gradients and V+A→P produced the lowest high gradient. The gradient fields were uneven throughout the ventricles. (Circulation 1992;85:1857–1864)

KEY WORDS • defibrillation • mapping, cardiac

The implantable defibrillator has been shown to decrease the incidence of sudden cardiac death in selected groups of patients known to be at high risk of recurrent ventricular fibrillation or tachycardia.1–3 The clinically available implantable defibrillator uses either a combination of transvenous catheter and epicardial patch electrodes or two epicardial patch electrodes.4 The widespread use of this device is hampered by the requirement of a major operative procedure for the implantation of the epicardial patch electrode. A technically less complicated approach without the need of a thoracotomy would reduce the risk of the implantation procedure and thus might allow the device to be used more readily for all patients who are at high risk of sudden death, including patients who previously might not have been candidates because of their poor surgical risks. Recently, ventricular defibrillation using nonthoracotomy electrode configurations has been tested in animals and humans, demonstrating the feasibility of this approach.5–8 The defibrillation efficacy with these nonthoracotomy electrode configurations is not as high as that with the conventional epicardial patch configuration.4,8 Thus, there is a need to improve the nonthoracotomy electrode configurations.

Current hypotheses about the mechanism of defibrillation postulate the need for a certain minimum potential gradient throughout all or most of the ventricular myocardium.9 It has been demonstrated that the potential gradient distributions on the epicardial surface produced by small epicardial patch electrodes are uneven, with a ratio of the highest to lowest gradient of 25:1.10 This unevenness of the gradient field contributes to the inefficiency of ventricular defibrillation. Defibrillation could probably be improved if electrode configurations could be developed that create a gradient field that is more even across the heart and in which the minimum gradient is higher for a given shock strength. Methods to determine the potential gradient distribution on the epicardial surface have been shown to be feasible.11,12 Recently, the technique has been extended to allow recordings of the potential gradient field in three dimensions.13 The purpose of this study was to use this improved technique to determine and compare the three-dimensional potential gradient field distributions created by four nonthoracotomy defibrillation electrode configurations.
configurations so that this information can be used to create electrode configurations with a more even gradient field, which may be more effective for ventricular defibrillation.

Methods

Experimental Procedure

Six mongrel dogs were anesthetized with pentobarbital (35 mg/kg body wt)\textsuperscript{11} followed by a continuous intravenous infusion of 0.05 mg/kg/min. Succinylcholine (1 mg/kg body wt) was injected intravenously on induction of anesthesia; supplemental doses of 0.25–0.5 mg/kg were given when required, but not more than once per hour. The dogs were intubated and ventilated with a Harvard respirator (Harvard Apparatus Company, South Natick, Mass.). The body temperature was monitored and maintained between 36.5 and 38°C by means of a warming blanket. Ringer’s lactate was infused continuously. Arterial blood gases and electrolytes, including calcium, were determined hourly. Potassium chloride, sodium bicarbonate, and calcium chloride were given when indicated. Through a median sternotomy, the heart was exposed and suspended in a pericardial cradle. An experimental defibrillation catheter (model R1068, Cardiac Pacemaker Incorporated, St. Paul, Minn.) was inserted through the right external jugular vein to the right ventricular apex. The catheter tip was anchored to the myocardium by protruding a stylet of the catheter through the myocardium to the epicardium and suturing the stylet to the epicardium. The catheter had four coiled, platinum-clad titanium electrodes. The distal electrode was 3 cm long, had a surface area of 2.8 cm\textsuperscript{2}, and was at the tip of the catheter, which was positioned and anchored to the right ventricular apex. The other three electrodes were each 1.5 cm long, had a surface area of 1.4 cm\textsuperscript{2}, and were spaced 3 cm apart. For this study, only the distal two electrodes were used. A cutaneous patch electrode (model 412, R2 Co., Morton Grove, Ill.) was placed on the left lower chest wall near the apex of the heart. Four shock electrode configurations were used: 1) V→A (Figure 1A); the right ventricular apical electrode (V) was used as cathode and the right atrial electrode (A) as anode. 2) V→P (Figure 1B); the right ventricular apical electrode was used as cathode and the cutaneous patch electrode (P) as anode. 3) V→A+P (Figure 1C); the right ventricular apical electrode was used as cathode. The right atrial electrode and the cutaneous patch electrode together were used as anode. 4) V+A→P (Figure 1D); the right ventricular apical electrode was combined with the right atrial electrode as cathode, and the cutaneous patch electrode was used as anode.

Recording electrodes were inserted into the atria, ventricles, and intraventricular septum to record from 128 sites as previously described.\textsuperscript{13} The needles were constructed of stainless steel covered with a coating of insulating epoxy resin. They were 0.9 mm in diameter. Thirty left ventricular plunge needles were used to record from subepicardial and subendocardial sites of the left ventricular free wall, with relatively equal spacing between needles. The recording sites were 1 mm and 9 mm from the epicardium. Six septal electrodes were inserted through the right ventricle and anchored within the intraventricular septum. Twenty-two right ventricular plunge needles were inserted to record from the subepicardium and subendocardium of
the right ventricular free wall, with the recording sites 1 and 4 mm from the epicardium. Eight atrial loop electrodes were sutured onto the epicardial surface of the atria just above the atrioventricular groove. Two loop electrodes were sutured onto the epicardial surface of the outflow tract of the right ventricle just below the conus. Two additional atrial electrodes were sutured onto the right high and left atria.

After the insertion of the recording electrodes, two chest tubes were placed: one at the posterior thorax for the drainage of fluid from the chest cavity and the other just below the sternum to evacuate air from the thoracic cavity. The sternotomy and incision were then sutured closed in layers and maintained airtight. A pair of pacing wires was sutured to the right atrial appendage for bipolar pacing at twice diastolic threshold. The heart was paced with 10 S, at a cycle length of 350 msec. At 300 msec after the last S, a 50-mA shock lasting 20 msec generated by a constant-current stimulator (Physio-Control Corp., Redmond, Wash.) was delivered through the above-mentioned electrode configurations. The shock potential distributions were recorded from the 128 sites simultaneously as previously described. All data were stored on videotape for off-line analysis. The voltage and current of the delivered shock were measured through a 5:1 voltage divider and 100Ω resistor, respectively, by a waveform analyzer (model Data 6000, Data Precision, Denver, Mass.) that digitized the signal at a frequency of 20 kHz. Average impedance of the shock was calculated from the digitized voltage and current. At the conclusion of the experiment, the heart was excised with the electrodes in place. The heart was cooled in ice/saline solution for 45 minutes. The plunge needles were then replaced by color-coded Teflon tubes. The atrial recording sites and the septal electrodes were marked by similar color-coded sutures. The heart was fixed in formalin for at least 2 days and then cut into slices approximately 2 mm thick as previously described. With a hand-held digitizer, the epicardial contour of each slice and the locations of the plunge needles were entered into a computer. The computer program then calculated the three-dimensional location of each recording electrode.

Data Processing and Analysis

The signals from the electrodes were entered into a computer-assisted mapping system capable of simultaneous recordings from 128 channels. The recordings were filtered from 0.1 to 500 Hz and digitized at a rate of 1,000 cycles/sec. Ten milliseconds before the shock, an external timing device signaled the mapping system to switch from bipolar to unipolar recording, to change the amplifier coupling to DC, and to modify the gain on each channel to a preset value appropriate for the shock. At the end of the shock, the external timing device signaled the mapping system to return to bipolar recordings and to the preexisting gain settings. The potential recorded from each channel was displayed on a computer terminal (Tektronix 4014, Tektronix, Inc., Beaverton, Ore.). On the basis of known calibration signals, a computer program then calculated the potential recorded by each channel at a single sample point near the midportion of the shock. The potential gradient (in volts per centimeter) was calculated by the method of Claydon. Because implantable defibrillator shocks are delivered by capacitors charged to a specified voltage, the potential gradients in this study were expressed per volt of shock strength by dividing the gradients by the voltage generated by the constant-current generator across the lead configurations. Heart displays (Figure 2) were constructed by computer programs that used the three-dimensional locations of the recording sites to produce a three-dimensional wireframe network for the subepicardium and subendocardium of the heart. Each recorded potential and calculated potential gradient was assigned to a polygon with the recording site at its center. The mass of ventricular myocardium between the subepicardial and the subendocardial polygons was divided in two. The epicardial half was assigned the potential gradient calculated for the subepicardial recording site. Similarly, the endocardial half was assigned the potential gradient calculated for the subendocardial recording site. These subdivisions were used to estimate the mass of ventricular myocardium reaching a given potential gradient for each shock (Figure 3).

For statistical analysis, two-way ANOVA was used for comparison among groups, and Duncan’s multiple range test was used to compare differences between groups. Results are presented as mean±SD. A value of p<0.05 was considered statistically significant for all analyses.

Results

Potential Gradient Fields

The potential distributions were measured for the shocks generated by each of the four original electrode configurations. The potential gradient fields were then calculated from the measured potentials and the electrode locations determined in three dimensions. Representative examples of the potential gradient fields from one animal are shown in Figure 2.

V→A. See Figure 2, panel A. For the configuration consisting only of electrodes on the defibrillation catheter, the potential gradient was highest at the lateral and posterior right ventricle, where the distal catheter shock electrode was located. The subendocardium, which was closer to the catheter electrode than was the subepicardium, had the higher gradients of the two. The potential gradient decreased rapidly with distance away from the catheter electrode such that the potential gradient was low throughout all of the left ventricular myocardium. The mean ratio of the highest to the lowest gradient with this electrode configuration for all dogs was 26.5:1 (Table 1), demonstrating the unevenness of the potential gradient distribution throughout the heart.

V→P. See Figure 2, panel B. The configuration using the distal catheter shock electrode as cathode and the cutaneous patch electrode as anode also produced a high-gradient area in the right ventricle close to the distal catheter electrode, similar to the V→A configuration. The potential gradient again decreased rapidly with distance from the catheter electrode, so the potential gradient over the left ventricular myocardium was relatively low. The mean ratio of highest to lowest
Figure 2. Facing page: Color computer-generated map of potential gradient field generated by the V→A electrode configuration in dog 6. Panels A–D show the fields for configurations V→A, V→P, V→A+P, and V→V→A+P, respectively. The epicardial surface of the heart is shown in the top two illustrations of each panel. The endocardial surface of the heart is shown in the bottom two illustrations of each panel. The two illustrations on the left show the anterior surface and the two illustrations on the right the posterior surface. The atria are not shown. The colors demonstrate the potential gradient distribution throughout the ventricular myocardium expressed in millivolts per centimeter per volt of shock. Panel E shows the outline of the heart with the major coronary arteries inserted for orientation in the other panels. LAD, left anterior descending coronary artery; RCA, right coronary artery; CX, circumflex artery; RV, right ventricle; LV, left ventricle. Panel F indicates the color scale used to represent the potential gradient values. See text for details.

Gradient was 19.1:1 (Table 1), indicating that this gradient field was also not even. V→A+P. See Figure 2, panel C. This configuration used the distal catheter electrode at the right ventricular apex as cathode, and the proximal catheter electrode at the right atrium was combined with the cutaneous patch electrode together as anode. The potential gradient was similar to that of the other configurations, except that the lowest gradients produced by this configuration were significantly larger than for the V→A configuration (Table 2) and that the highest gradients produced by this configuration were significantly higher than for the V→P and the V+P configurations (Table 3). The ratio of highest to lowest gradient was 17.6:1 (Table 1). The impedance for this configuration was less than for the other configurations (Table 4).

V→A→P. See Figure 2, panel D. This configuration used the distal catheter electrode at the right ventricular apex and the more proximal catheter electrode at the right atrium together as cathode and the cutaneous patch electrode as anode. The potential gradient was again highest in the right ventricle, where the distal catheter electrode was located, but the highest gradients were significantly lower than those for the V→A and V→A+P configurations (Table 3). The potential gradient over the left ventricle was low, particularly in the basal left ventricle. This configuration again produced an uneven gradient field, with a ratio of highest to lowest gradient of 16.5:1 (Table 1).

Myocardial Mass–Potential Gradient Curves

The potential gradients produced by the shocks were estimated for each myocardial region of the heart as described in “Methods.” For each electrode configuration for each dog, the percentage of the total ventricular mass less than a given potential gradient was plotted against that potential gradient. Figure 3 shows the plots for the electrode configurations for each dog. The lower portion of the curves represents the percent of the ventricular mass exposed to a low potential gradient.

Figure 3. Graphs showing percent of ventricular mass with a potential gradient less than or equal to that indicated on the abscissa. Panels A–F have four curves, each plotting the data for each of the four electrode configurations. The more vertical the line, the more even the potential gradient field. When a minimum potential gradient is required throughout the ventricles for defibrillation, the electrode configuration for which the bottom portion of the curve is farthest to the right will defibrillate with the smallest voltage shock.
For example, for dog 1 (Figure 3A), 10% of the ventricular myocardial mass had a potential gradient <4 mV/cm/V for the $V+A\rightarrow P$ configuration, whereas for the $V\rightarrow A+P$ configuration, 10% of the ventricular myocardial mass had a potential gradient <8 mV/cm/V. At the 10% level, the $V+A\rightarrow P$ curves for all six dogs were either the farthest to the right or among the rightmost of the four curves (Table 2). This suggests that for the same shock strength, the $V+A\rightarrow P$ configuration produced a smaller region of low potential gradient than did the other three configurations.

The top part of the curves represents potential gradients for which most of the ventricular myocardial mass was exposed to a smaller or equal gradient. For example, for dog 1 (Figure 3A), 90% of the ventricular myocardium had a potential gradient $\leq 11$ mV/cm/V for the $V+A\rightarrow P$ configuration. As a corollary, 10% of the ventricular myocardium had a potential gradient $>11$ mV/cm/V. In comparison, 90% of the ventricular myocardium had a potential gradient $\leq 24$ mV/cm/V for the $V\rightarrow A+P$ configuration, i.e., 10% of the myocardium had a gradient $>24$ mV/cm/V. In all six dogs, the top parts of the $V\rightarrow A+P$ curves are to the right of the other three curves (Table 3), suggesting that although the $V+A\rightarrow P$ configuration was able to increase the potential gradient at the low-gradient part of the curve, it also increased the potential gradient at the high-gradient part of the curve.

### Table 1. Highest and Lowest Potential Gradients Produced by the Four Electrode Configurations

<table>
<thead>
<tr>
<th>Electrode configurations</th>
<th>Dog</th>
<th>$V\rightarrow A$</th>
<th>$V\rightarrow P$</th>
<th>$V\rightarrow A+P$</th>
<th>$V+A\rightarrow P$</th>
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<td>27.7/1.8</td>
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<td>2</td>
<td>96.8/3.1</td>
<td>77.1/4.1</td>
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<td>54.3/1.7</td>
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<td></td>
<td>3</td>
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<td>112.2/3.4</td>
<td>168.8/7.8</td>
<td>66.6/3.7</td>
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<td>35.7/2.2</td>
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<td>60.4/2.6</td>
<td>80.4/5.7</td>
<td>91.2/5.3</td>
<td>55.5/4.3</td>
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<tr>
<td></td>
<td>6</td>
<td>60.7/3.2</td>
<td>66.7/5.1</td>
<td>81.9/5.0</td>
<td>55.0/6.7</td>
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<td>Mean (H/L)</td>
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<td>Mean (ratio)</td>
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<td>7.2</td>
<td>2.6</td>
<td>8.2</td>
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</table>

Values are in millivolts per centimeter per volt. $V\rightarrow A$, $V\rightarrow P$, $V\rightarrow A+P$, $V+A\rightarrow P$, four electrode configurations tested in these studies; see text for description. H, highest potential gradient produced by a 50-mA shock through the electrodes; L, lowest potential gradient produced by the same shock; ratio is the ratio of highest to lowest gradient produced by the shock.

### Discussion

The automatic implantable cardioverter/defibrillator has revolutionized the management of patients with malignant ventricular arrhythmias.\(^1\)\(^-\)\(^3\) Despite its wide use to terminate ventricular fibrillation, little is known about the distribution of the electric fields throughout the heart produced by defibrillation shocks. Extracellular current density and potential gradient are the variables thought to correlate most closely with changes in transmembrane potentials caused by the shock and hence with the ability to defibrillate.\(^20\) Current density and potential gradient are closely related to each other by the conductivity of the tissue.

The measurement of local potentials at multiple sites in the heart and the knowledge of the locations where the local potentials are sampled allow the estimation of the change of potential over distance (the potential gradient). A few years ago, these measurements were obtained in a few sites within the heart.\(^21\)\(^,\)\(^22\) Computer-assisted multichannel mapping now allows many more potentials to be recorded simultaneously. With this technology, the epicardial potential distribution caused by shocks delivered from epicardial shock electrodes\(^10\)\(^-\)\(^12\) has been recorded over the heart. Although these studies provide important information about the mechanism of ventricular defibrillation, they lack the determination of transmural potential gradients, which may be the major component when shocks are

### Table 2. Tenth Percentile of Potential Gradient in the Ventricles

<table>
<thead>
<tr>
<th>Electrode configurations</th>
<th>Dog</th>
<th>$V\rightarrow A$</th>
<th>$V\rightarrow P$</th>
<th>$V\rightarrow A+P$</th>
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Ten percent of the ventricular myocardium had a potential gradient (mV/cm/V) less than that presented in the table.

* $V\rightarrow A+P$ significantly higher than $V\rightarrow A$.

### Table 3. Ninetieth Percentile of Potential Gradient in the Ventricles

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<th>Electrode configurations</th>
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Ninety percent of the ventricular myocardium had a potential gradient (mV/cm/V) less than that presented in the table.

* $V\rightarrow P$ significantly lower than $V\rightarrow A+P$.
† $V+A\rightarrow P$ significantly lower than $V\rightarrow A$ and $V\rightarrow A+P$. 

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\(^1\) Edvardsen, J. et al. \(\text{Circulation}\) 1982, \text{No} 5, \text{pp} 55-64.
\(^2\) Edvardsen, J. et al. \(\text{Circulation}\) 1982, \text{No} 5, \text{pp} 55-64.
\(^3\) Edvardsen, J. et al. \(\text{Circulation}\) 1982, \text{No} 5, \text{pp} 55-64.
delivered through a transvenous catheter. To overcome this limitation, the multichannel mapping technique was improved to allow recording of potentials and calculation of potential gradients in three dimensions throughout the ventricular myocardium. We used this technique to estimate the gradient fields of four shock electrode configurations that have been reported in the literature.

When the results of this study are evaluated, it should be kept in mind that there are probably large errors in the calculated gradients. One source of error is the error in the recorded potentials and electrode locations. Another source of error is the assumption used by the numerical algorithm for estimating the potential gradient at each electrode that the gradient is constant between the electrode and its neighbors. This assumption should cause the greatest errors in the regions of high gradient near the catheter defibrillation electrodes, where the gradients are changing the most rapidly. These considerations suggest that the low and particularly the high gradient values given in Table 1 are subject to substantial error. The fact that the gradient fields are so uneven for the different configurations, however, suggests that the identification of the regions of high and low gradient are probably subject to less error than are the absolute values of the gradients.

The configuration in which current is directed from the right ventricular apical catheter electrode to both the right atrial catheter electrode and the cutaneous patch produced higher gradients throughout the heart and shifted the myocardial mass-potential gradient curve to the right compared with the other three configurations (Figure 3). A possible explanation for this favorable potential gradient distribution is that the two anodal sources are away from the ventricular myocardium, so that any low-voltage gradient area generated by electric field cancellation between the two anodes may fall outside the myocardium. This configuration may therefore have the advantage that a lower total shock strength may be required to reach a certain minimal gradient over most or all of the ventricular myocardium for successful defibrillation. Most hypotheses about the mechanism of ventricular defibrillation suggest that a certain minimal potential gradient is required throughout most or all of the ventricular myocardium. Therefore, this configuration may be the best of the four for defibrillation, as has been suggested by clinical studies.

The large combined surface area of the two anodes of an orthogonal electrode configuration decreased impedance to the shock, so more current flowed per volt of shock (Table 4). However, this current all had to pass through the single right ventricular apical cathode. Thus, the high-gradient zone close to the right ventricular apical electrode produced by this configuration had a higher gradient than those of the other three configurations. There is evidence to suggest that excessively high potential gradients may have deleterious effects, including conduction abnormalities, arrhythmias, and even ventricular fibrillation. In this respect, the V→A→P configuration, in which the two electrodes on the catheter were combined as the cathode, may be advantageous, because the gradient at the high-gradient area was lower than that for the other three configurations.

In conclusion, this study demonstrates that 1) it is possible to estimate shock potential gradient fields in three dimensions, 2) the V→A→P configuration produced higher gradients throughout the heart and shifted the myocardial mass-potential gradient curve to the right, 3) the high gradient area for the V→A→P configuration was lower than that for the other configurations, and 4) all four configurations produced uneven gradient fields.

### TABLE 4. Impedances of the Four Configurations

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*V→A+P significantly lower than the other configurations.

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