Cardiac Potential and Potential Gradient Fields Generated by Single, Combined, and Sequential Shocks During Ventricular Defibrillation

J. Marcus Wharton, MD; Patrick D. Wolf, MS; William M. Smith, PhD; Peng-Sheng Chen, MD; David W. Frazier, MD; Seitaro Yabe, MD; Ned Danieleys, MD; and Raymond E. Ideker, MD, PhD

Background. Potential gradient field determination may be a helpful means of describing the effects of defibrillation shocks; however, potential gradient field requirements for defibrillation with different electrode configurations have not been established.

Methods and Results. To evaluate the field requirements for defibrillation, potential fields during defibrillation shocks and the following ventricular activations were recorded with 74 epicardial electrodes in 12 open-chest dogs with the use of a computerized mapping system. Shock electrodes (2.64 cm²) were attached to the lateral right atrium (R), lateral left ventricular base (L), and left ventricular apex (V). Four electrode configurations were tested: single shocks of 14-msec duration given to two single anode—single cathode configurations, R:V and L:V, and to one dual anode—single cathode configuration, (R+L):V; and sequential 7-msec shocks separated by 1 msec given to R:V and L:V (R:V→L:V). Defibrillation threshold (DFT) current was significantly lower for R:V→L:V than for the other configurations and markedly higher for L:V. Despite these differences, the minimum potential gradients measured at DFT were not significantly different (approximately 6–7 V/cm for each electrode configuration). Potential gradient fields generated by the electrode configurations were markedly uneven, with a 15–27-fold change from lowest to highest gradient, with the greatest decrease in gradient occurring near the shock electrodes. Although gradient fields varied with the electrode configuration, all configurations produced weak fields along the right ventricular base. Early sites of epicardial activation after all unsuccessful shocks occurred in areas in which the field was weak; 87% occurred at sites with gradients less than 15 V/cm. Ventricular tachycardia originating in high gradient areas near shock electrodes followed 11 of 67 successful shocks.

Conclusions. These data suggest that 1) defibrillation fields created by small epicardial electrodes are very uneven; 2) achievement of a certain minimum potential gradient over both ventricles is necessary for ventricular defibrillation; 3) the difference in shock strength required to achieve this minimum gradient over both ventricles may explain the differences in DFTs for various electrode configurations; and 4) high gradient areas in the uneven fields can induce ectopic activation after successful shocks. (Circulation 1992;85:1510–1523)

KEY WORDS • defibrillation • fields, gradient

To defibrillate with the least amount of energy, the potential field generated by the defibrillating shock must be optimized. The potential gradient has been reported to be closely related to the success or failure of defibrillation.1 If a certain minimum potential gradient is necessary over the myocardium to achieve defibrillation, shocks that do not produce this minimum potential gradient will not be successful. Furthermore, the defibrillation threshold (DFT) should be determined by the shock strength required to obtain this minimum gradient across the myocardium; thus, potential gradients may explain the relative efficacy of different electrode configurations. Though a few studies have measured potential gradients and gradient fields during electrical shocks in sinus rhythm or ventricular fibrillation,2–5 systematic evaluation of field requirements for multiple electrode configurations to determine the minimum magnitude and spatial requirements of potential gradient fields has not been performed. In addition, mapping of potential gradient fields may help to explain the increased efficacy of some sequential shock electrode configurations. The

All editorial decisions for this article, including selection of reviewers and the final decision, were made by a guest editor. This procedure applies to all manuscripts with authors from the University of California San Diego or UCSD Medical Center.

From the Departments of Medicine and Pathology (J.M.W., P.D.W., W.M.S., D.W.F., S.Y., N.D., R.E.I.), Duke University Medical Center, Durham, N.C., and the Cardiology Division (P.S.C.), University of California Medical Center, San Diego, Calif.

Supported in part by National Institutes of Health research grants HL-17670, HL-28429, HL-33637, HL-07063, HL-44066, and HL-42760; American Heart Association North Carolina Affiliate grant-in-aid 1985–86 A-06; and a grant from the Fannie E. Ripple Foundation. J.M.W. received the North American Society of Pacing and Electrophysiology Young Investigator Award for a presentation of a preliminary version of this article.

Address for correspondence: J. Marcus Wharton, MD, PO Box 3816, Duke University Medical Center, Durham, NC 27710.

Received January 22, 1991; revision accepted November 21, 199
present study was designed to 1) describe the potential and potential gradient fields generated by electrode configurations with different defibrillation efficacies; 2) correlate the field characteristics with the efficacy of each electrode configuration; 3) describe the field characteristics necessary for defibrillation; and 4) define further the relation of field strength to the site of initial ventricular activation after successful and unsuccessful defibrillation shocks.

Methods

Surgical Preparation

In 12 mongrel dogs (mean weight, 19.7±4.7 kg), anesthesia was induced with intravenous pentobarbital 30–35 mg/kg and maintained with a dose of 0.05 mg/kg/min. Skeletal muscle paralysis was obtained with intravenous succinylcholine 1 mg/kg and maintained with a dose of 0.25–0.50 mg/kg/hr as needed to prevent skeletal muscle contraction. The dogs were intubated and mechanically ventilated (model 607, Harvard Apparatus Company, Inc.). A midline sternotomy was performed, and the heart was suspended in a pericardial cradle. The nylon sock, which was placed over the ventricles for mapping purposes (see below), helped prevent desiccation of the exposed portions of the heart; warm saline was also applied repeatedly to the sock to keep it moist. Warm lactated Ringer's solution was infused continuously through a peripheral intravenous cannula, and blood pressure was monitored through a femoral artery cannula connected to a transducer (model 1280, Hewlett-Packard Co.). Arterial blood was sampled every 30–60 minutes (or more frequently if indicated) and pH, Pco2, Po2, HCO3-, Na+, K+, and Ca++ levels were monitored and corrected as necessary to maintain these parameters within normal limits. Core temperature was monitored with a tongue thermometer and maintained at 37°C using a heating blanket. A heating lamp warmed the open chest. All studies were performed in accordance with the guidelines for the use of experimental animals established by the American Physiological Society.

Recording and Shock Electrodes

A nylon sock with 72 evenly spaced button electrodes was applied over the heart and secured to the atria, ativoventricular groove fat, or pericardium to ensure stability of location. The button electrodes were placed in seven rows from apex to base (Figure 1A), with the basal row extending to the ativoventricular groove. Each button encases two electrodes with centers 1.5 mm apart (1.0 mm edge to edge). Unipolar or bipolar recordings were obtained from these electrodes depending on the setting of the mapping system (see below). A clamp electrode attached to the left leg served as the reference for unipolar recordings.

Three solid circular titanium mesh electrodes with a diameter of 2.0 cm (total surface area, 3.14 cm²) and with a bipolar button recording electrode (surface area of plastic button, 0.50 cm²) centered underneath were used to deliver shocks (Figures 1A and 1B). The total area of the shock electrode exposed to myocardium and not bounded by insulator was a circular ring with a width of 0.6 cm (surface area, 2.64 cm²). Shock electrodes were sewn securely to the epicardium of the lateral wall of the right atrium near the junction of the superior vena cava (electrode R in Figures 1A and 1B), the lateral free wall of the left ventricle adjacent to the ativoventricular groove (electrode L), and the left ventricular apex (electrode V). The location of the L and V electrodes was varied slightly to avoid placement over or near coronary arteries. Two pairs of stainless steel wires were inserted into the right ventricular free wall approximately 1 cm apart for ventricular pacing and sensing.

Defibrillation Protocol

After induction of ventricular fibrillation with a short burst of 60-Hz alternating current in the open-chest preparation, single truncated exponential defibrillation shocks with a duration of 14 msec were given with a specially designed 750-microfarad defibrillator (Intermedics, Inc.) through one of three anode–cathode electrode configurations, abbreviated R:V, L:V, and (R+L):V. The last abbreviation indicates that the R and L electrodes are combined as a common anode. One sequential shock (abbreviated R:V→L:V) was tested in which a 7-msec R:V was followed by a 7-msec L:V separated by a pause of 1 msec. The 14-msec monophasic waveform had a 16% tilt when delivered across a test resistor of 10 Ω.

DFTs were determined using a modified Purdue method. The initial shocks were given at an estimated strength of 4 J (2 J to each shock for the sequential pair) to each of the electrode configurations in random order. The shock energy to be delivered was estimated by measuring the current during a 100-V shock delivered in sinus rhythm from an electrode configuration, calculating the impedance, and then calculating the voltage necessary to deliver 4 J (or 2 J sequentially) assuming no change in impedance. Because impedances for R:V and L:V were different, potentials required to deliver 2 J to each component of the sequential shock were not the same. Each shock was given approximately 15 seconds after the induction of ventricular fibrillation; when it was unsuccessful, a 10–30-J rescue shock (defibrillator model 7802B, Hewlett-Packard) was given through hand-held paddles placed on the pericardium over the right and left ventricles. Temporary pacing and/or brief cardiac massage was given during periods of asystole after cardioversion. Another shock was not attempted for at least 5–10 minutes or until baseline heart rate and blood pressure were reattained.

When a defibrillation attempt failed, the energy level for the subsequent shock was increased by 20% above the delivered energy from the preceding unsuccessful shock for that electrode configuration, and defibrillation was reattempted. Estimation of the voltage required to deliver the desired energy for each electrode configuration was based on the impedance measured from the preceding shock with the same electrode configuration. When a defibrillation shock was successful, the energy for the next shock for that electrode configuration was decreased by 10% until defibrillation was not obtained. Each electrode configuration was tested in random order for each incremental change in energy. The lowest energy resulting in defibrillation was called the DFT.

Electrode Recordings

Before and after each attempted defibrillation, epicardial activation was recorded by the 75 bipolar button
electrodes (74 of which were on the ventricles). Amplifier gains were set for each channel for optimal signal recording. Electrogram signals were filtered to pass 0.1–500 Hz before and after attempted defibrillation. Ten milliseconds before a shock, recording was switched to unipolar at a second set of gains (previously determined to be appropriate for the voltage of the shock to be delivered) with a low-pass filter of 500 Hz and high-pass filter direct-current coupled (Figure 1C). Ten milliseconds allowed adequate time to establish a baseline for measuring potentials generated by the shocks. A voltage attenuator produced by a 1-GΩ resistor at the front end of the amplifier was simultaneously used with the other changes to reduce the electrode potential by a factor of 1,000. During the shock, epicardial potentials were measured by the 75 button electrodes, which included the three electrodes located beneath the shock electrodes. The potential delivered to the shock electrodes was also directly measured. Current delivered to each electrode pair for single, combined, and sequential electrode configurations was measured on an oscilloscope. The voltage attenuator was switched off 1 msec after the shock, and recording of bipolar electrograms was resumed (Figure 1C).
After completion of the study, the dog was killed by induction of ventricular fibrillation. The button and shock electrode locations were marked with color-coded pins. Their locations were subsequently transcribed to a computer-generated, three-dimensional display of the dog heart for generation of surface maps.

Data Analysis

Data were digitized at 1,000 Hz and recorded on videocassette for subsequent analysis. Shock potentials at the median time of the shock were determined for every recording electrode by a computer program with known calibration signals. Total delivered potential was calculated from the measured shock electrode potentials as the anodal potential or mean anodal potential for (R+L):V, minus the cathodal potential.

Total energy delivered was calculated for all dogs from the equation

\[ J = \text{VIT} \]

where \( J \) is energy in joules, \( \text{V} \) is mean delivered potential in volts, \( I \) is mean current in amperes, and \( t \) is the duration of the shock in seconds. In five dogs, a digital oscilloscope was used to store the measured current and potential waveforms, and the power curve was integrated over time to obtain the delivered energy. Energy measurements by this means were nearly identical to those made with the above equation (\( r=0.9998; \) slope, 0.99).

Impedance was calculated from Ohm’s law, assuming a primarily resistive element to impedance in the range of potentials used in this study (see “Appendix”):

\[ R = \Delta E/I \]

For total impedance, \( \Delta E \) equals the total delivered potential. For electrode–myocardial interface impedance, \( \Delta E \) is the potential difference from shock electrode to the button electrode underneath it. Cardiac impedance was calculated as the difference between total and electrode–myocardial interface impedance. Because intracardiac and shunt current were not measured in the parallel circuit of (R+L):V, direct measurement of the associated cardiac impedance was not possible; it was estimated by averaging the potentials measured underneath the anodal shock electrodes and assuming a single circuit impedance.

Potential gradients were calculated as described previously. Briefly, the nearest neighbors of each recording electrode were determined, and the interelectrode distance of each electrode to its neighbors was calculated from their x-y-z coordinates. From the potential differences and interelectrode distances between an electrode and each of its nearest neighbors, a method that minimized the summed squared error was used to estimate the potential gradient at the recording electrode. The minimum and maximum gradient for each defibrillation electrode configuration for each dog was determined from the lowest and highest gradient, respectively, from any recording electrode for a given shock.

Postshock epicardial activation times at each recording electrode, referenced to the onset of the defibrillation shock, were determined by using the peak of monophasic waveforms or the maximum potential change of biphasic or multiphasic waveforms as the point of activation. The earliest site of postshock ventricular activation was defined as the recording electrode site that first recorded activation after the shock. Other early sites were defined as electrode sites with activation times earlier than surrounding neighbor electrodes. Epicardial activation times, potentials, and potential gradients were projected onto two-dimensional representations of the dog heart for visual analysis.

Data are expressed as mean±SD. Repeated-measures analysis of variance was performed to compare measured and calculated values between the multiple electrode configurations. Pearson correlation coefficients were used to compare different means of energy calculation and to compare potential fields to test the rule of superposition. A probability value less than 0.05 was considered significant.

<table>
<thead>
<tr>
<th>Table 1. Global and Field Parameters at Defibrillation Threshold</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
</tr>
<tr>
<td>Global parameters</td>
</tr>
<tr>
<td>Voltage (V)</td>
</tr>
<tr>
<td>Current (A)</td>
</tr>
<tr>
<td>Energy (J)</td>
</tr>
<tr>
<td>Field parameters</td>
</tr>
<tr>
<td>Minimum gradient (V/cm)</td>
</tr>
<tr>
<td>Maximum gradient (V/cm)</td>
</tr>
</tbody>
</table>

R:V and L:V, single anode-single cathode configurations; (R+L):V, dual anode-single cathode configuration; R:V→L:V, sequential θ-msec shocks separated by 1 msec given to R:V and L:V.

†Represents field parameter from maximum combined gradient field (see text).
significantly different from \( R:V \) and \( (R+L):V \). \( R:V \) and \( (R+L):V \) were similar in terms of energy and voltage, but \( (R+L):V \) required significantly more current than \( R:V \) \( (p<0.05) \). On the other hand, \( L:V \) was markedly worse in all parameters. At DFT, mean voltage and current for \( L:V \) were both approximately 1.5 times higher and energy was approximately three times higher than for the other electrode configurations \( (p \leq 0.005 \) for all parameters).

There were 329 shocks delivered: 81 \( R:V \), 114 \( L:V \), 80 \( (R+L):V \), and 54 \( R:V \rightarrow L:V \). The mean numbers of shocks necessary to obtain DFT for each of these electrode configurations were \( 7 \pm 2 \), \( 10 \pm 2 \), \( 7 \pm 3 \), and \( 5 \pm 1 \), respectively. Determination of the DFT for \( L:V \) required a significantly greater number of shocks than for the other electrode configurations.

Potential and Potential Gradient Fields

The potential fields, potential gradient fields, and the postshock activation sequences were mapped for all unsuccessful and successful shocks. For each electrode configuration, the potential and potential gradient fields generated were similar among different experimental animals. A representative example of maps of the potential field and potential gradient field for each component of the sequential shock configuration is shown in Figure 2. Each component shock of \( R:V \rightarrow L:V \) was similar to \( R:V \) and \( L:V \) shocks of equivalent magnitude given singularly. Potential measured at the recording electrode immediately underneath the shock electrode was approximately two thirds of the measured delivered potential for both components of the sequential shock. Mean potential decrease at the electrode-myocardial interface ranged from 27% to 29% of the applied potential for all of the electrode configurations tested. Decline in potential across the myocardium was relatively symmetrical with respect to the interelectrode axis of each configuration and was most rapid near the shock electrodes. As can be seen in Figures 2A and 2B, potential change was small across the base of the right ventricle for both \( R:V \) and \( L:V \). Potential change was also small across the entire base of the right ventricle with \( R:V \) shocks and across the posterior left ventricular base with \( L:V \) shocks.

When the combined anodal configuration \( (R+L):V \) was used, the potential field was similar to \( R:V \) except for a rapidly decreasing component which rapidly decreased with distance from the left ventricular anode (Figure 3A). From the rule of superposition, the potential field generated by \( (R+L):V \) should be equal to the sum of the potential fields generated by \( R:V \) and \( L:V \) if current flow were equal between each component of \( (R+L):V \) and the corresponding \( R:V \) and \( L:V \) shocks. Equal currents \( (\pm 0.1 \) A) among all necessary component shocks were available for four shock strengths in two dogs. The correlation between the actually measured \( (R+L):V \) potential field and the field calculated from superpositioning equicurrent \( R:V \) and \( L:V \) fields for these four shocks was excellent \( (r=0.993-0.998) \).

The small size of the shock electrodes resulted in very inhomogeneous potential gradient fields (Figures 2C and 2D). The unevenness of the potential gradient fields for each electrode configuration is quantitated by comparison of minimum and maximum gradients at DFT in Table 1. There was a 19.5±13.2-, 27.6±17.7-, 23.3±15.3-, and 15.7±4.4-fold change between the minimum and maximum measured potential gradient at the DFT for \( R:V \), \( L:V \), \( (R+L):V \), and the composite field (see below) of \( (R:V) \rightarrow (L:V) \), respectively. Paralleling the changes seen in the potential fields, potential gradient decline was most rapid near the shock electrodes and more gradual at sites more distant from the shock electrodes. Relatively weak potential gradient fields (less than 10 V/cm) were produced across the base of the right ventricle by all electrode configurations used. \( R:V \) also generated weak fields across most of the base of the left ventricle and \( L:V \) across the posterior base of the left ventricle. For \( (R+L):V \), the gradient field was less than 10 V/cm across the base of both ventricles except for a relatively small area surrounding the \( L \) electrode (Figure 3B). Except for the area enhanced by the \( L \) anode, the gradient field was not greatly different from that of \( R:V \), which may explain their similar DFTs. Figure 4 graphically illustrates the distribution of low gradient areas for each electrode configuration.

As can be anticipated from the individual gradient fields for \( R:V \) and \( L:V \), neither component of the sequential shocks obtained high gradients across the right ventricular base or across portions of the base of the left ventricle anteriorly and posteriorly near the right ventricle and distant from the \( L \) electrode. To illustrate this, composite maps were generated by using the maximum gradient measured from either of the two component shocks at each electrode site (Figure 2E). This is justifiable because both 7- and 14-msec truncated exponential shocks occur on the relatively flat portion of the strength–duration curve\(^{10} \) and presumably should have approximately the same potential gradient field requirements for defibrillation. Such composite maps demonstrated that neither of the sequential shocks created potential gradients greater than 10 V/cm across the base of the right ventricle. In some experiments, gradients greater than 10 V/cm were also not created by the sequential shocks across the anterior and/or posterior base of the left ventricle. However, 52±19% of recording electrodes with gradients less than or equal to 10 V/cm during the \( R:V \) component of the sequential shock at DFT recorded gradients greater than 10 V/cm during the \( L:V \) component. The minimum gradient for each component of the sequential shock at DFT was 5.3±0.8 V/cm and 4.4±0.9 V/cm, respectively. These were significantly less \( (p=0.012) \) than the minimum gradient of 6.6±1.8 V/cm for the composite field. The ability of two sequential fields of lower minimum gradient to generate a composite field with higher minimum gradient presumably accounts for the greater efficacy of the sequential shock compared with either component alone. The minimum gradients at DFT for the \( R:V \) and \( L:V \) components of the sequential shock were significantly less \( (p<0.01) \) than for those at DFT for the respective single 14-msec shocks. However, the minimum gradients at DFT for the composite field of the sequential shock and for the fields of the single \( R:V \) and \( L:V \) shocks were similar.

Minimum Gradient and Defibrillation Parameters

\( R:V \rightarrow L:V \) required significantly less and \( L:V \) required significantly more current to defibrillate than
FIGURE 2. Maps of potential fields for sequential 208-V R:V (two single anode–single cathode configuration) (panel A) and 205-V L:V (two single anode–single cathode configuration) (panel B) shocks are illustrated for an unsuccessful defibrillation attempt. Left-hand figure of each panel represents anterior half and right-hand the posterior half of the ventricles. Numbers indicate location of recording epicardial electrodes and potential in volts (relative to the left leg) recorded at each site. Boxed-in numbers represent measured potential delivered to shock electrodes: number at upper left-hand corner in panel A represents the anode at the right atrium (R); number at left base in panel B represents the left basal anode (L); number at left-hand apex of both panels A and B represent the left ventricular apical cathode (V). Numbers immediately beneath the right atrial or to the left of the left basal anodal potentials and above the cathodal potentials indicate potentials measured by electrode immediately beneath shock electrodes. Isopotential lines are separated by 20 V. Note relative symmetry of isopotential lines; they approximately describe circles perpendicular to the long axis of the heart during the R:V shock. During the L:V shock, there is also relative symmetry of the potential field along the axis between L and V. Fields generated between each 7-msec duration component of the sequential shock were similar to the corresponding 14-msec single shock of similar delivered potential. Panels C and D: Derived potential gradient fields for R:V and L:V components of sequential shock illustrated above are shown. Numbers represent potential gradient at each site in V/cm. Lowest gradients (≤10 V/cm) are enclosed in cross-hatched area. Isogradient lines are separated by 20 V/cm and parallel the symmetry of potential fields. During R:V shock (panel C), gradient at electrode beneath apical cathode was 48 V/cm. During the L:V shock (panel D), gradient at electrode underneath left ventricular basal anode was 48 V/cm and underneath apical cathode was 43 V/cm. Panel E: To observe the effect of sequential shocks on improving gradient distribution across the epicardium, maximum gradient from either shock at each electrode site was displayed. "Maximum combined gradient field" for sequential shock illustrated in panels C and D is shown in panel E. Note that extent of myocardium not obtaining a gradient greater than 10 V/cm is substantially reduced compared with gradient fields for component shocks. Panel F: Map of epicardial activation after unsuccessful sequential shock. Numbers represent times in milliseconds when epicardial activation occurs at the recording electrode, with time zero being the start of the shock. Filled circles represent sites of electrodes in which adequate bipolar recording of activation was not obtained. Isochronal lines are separated by 20 msec. Epicardial activation map shows three sites of early epicardial activation arising along the atrioventricular groove in which neither shock electrode generated a strong field. Thick arrow indicates site of earliest postshock epicardial activation; thin arrows indicate other somewhat later sites of epicardial activation. Subsequently, there is rather uniform spread across the ventricles from these foci.
A

\[ (R+L):V \]

\[ +35 \]

RV LV LV RV

\[ +18 \]

-184

POTENTIAL FIELD

B

GRADIENT FIELD

C

POST-SHOCK ACTIVATION

FIGURE 3. Maps. Panel A: Potential field for 220-V subthreshold shock with the \((R+L):V\) (dual anode–single cathode) electrode configuration from the same experimental animal as in Figure 2. Potential delivered to right atrial anode was \(+35\) V (upper left-hand corner of figure), to the left ventricular anode, \(+36\) V, and to the apical cathode, \(-184\) V. Potentials measured underneath respective shock electrodes were \(+18\), \(+21\), and \(-124\) V. Potential drop across anodes was similar. Potential drops rapidly around the L shock electrode with little distortion otherwise of the field generated by \(R:V\) (single anode–single cathode configuration) (Figure 2A). Isopotential lines are thus relatively symmetrical across apical half of myocardium. Panel B: Gradient field for the same \((R+L):V\) shock. Largest gradient (82 V/cm) is located beneath apical cathode. Gradient measured underneath left ventricular anode was 34 V/cm. Gradient field is weak across most of the base of the ventricles except for area immediately surrounding the L electrode. Panel C: Map of epicardial activity after shock illustrated in panel A reveals two electrodes recording the same postshock activation time of 63 msec, presumably resulting from an area of earlier activation at a site between the two recording electrodes (thick arrow) propagating out to recording electrodes at approximately the same time. Two other early sites (thin arrows) of postshock activation are seen, one at the base of the left ventricle and the other along the right ventricular free wall, both in areas in which gradient field is weak.

R:V or (R+L):V (Table 1). In addition, the potential fields generated were different for each electrode configuration. Despite these differences, the minimum potential gradients measured at DFT for each of these electrode configurations were approximately 6–7 V/cm and were not significantly different (Table 1). This relation is also true for the composite field of the sequential shock configuration; however, the individual component shock fields had minimum gradients that were somewhat lower (see above). The mean maximum gradient for L:V was significantly larger than for the other electrode configurations \((p<0.009)\), although the latter were not different from each other (Table 1). Thus, despite the differences in electrode configuration, in generated potential fields, and in voltage, current and energy required for defibrillation, there was not a significant difference in the minimum potential gradient generated by any of the electrode configurations at their DFTs. This implies that there is a minimum potential gradient that must be obtained over both ventricles for electrical defibrillation in normal myocardium. For sequential shocks, this minimum gradient must be obtained by at least one of the two shocks at each ventricular site.

Sites of Epicardial Activation After Subthreshold Shocks

There were 262 unsuccessful shocks, with an average of 2.3±1.0 early epicardial sites per shock. The number of early sites did not differ significantly with the electrode configuration used, as can be seen in Table 2. Initial activation occurred where gradients were weak for each electrode configuration (Figures 2A and 3C). In general, early sites occurred where the gradients were less than 10 V/cm (Figure 4). There was no evidence of early epicardial activation occurring in high gradient areas after unsuccessful shocks. The distribution of potential gradients at the sites of early activation for all unsuccessful shocks was approximately normal with a skew to the left (Figure 5). The mean potential gradient at sites of early activation after unsuccessful shocks was 9.5±5.4 V/cm; median was 7.9 V/cm. Ninety-eight percent had gradients less than 25 V/cm, and 87% had gradients less than or equal to 15 V/cm. Fifty-two percent of the earliest sites were within 1 V/cm of the minimum measured potential gradient for the corresponding shock. The largest discrepancy between minimum measured gradient and the gradient at the site of earliest epicardial activation was 21.6 V/cm (7.7 and 29.3 V/cm, respectively), occurring with an unsuccessful sequential shock. Mean gradient at the sites of early activation for each electrode configuration after the highest subthreshold shocks is shown in Table 2.

Sites of Epicardial Activation After Successful Shocks

Of the 67 successful shocks, 11 (16%) were followed by the immediate resumption of supraventricular rhythm (type A defibrillation) and 55 (83%) were followed by one to several ventricular ectopic complexes before the resumption of supraventricular rhythm (type B defibrillation).\(^3\)\(^,\)\(^8\)\(^,\)\(^9\) Type A defibrillation occurred in six of 19 R:V shocks, none of 14 L:V shocks, four of 18 (R+L):V shocks, and one of 16 R:→L:V shocks. Of the 55 defibrillations followed by transient ventricular
ectopic activity, there were 94 early sites or a mean of 1.7±0.8 per successful shock. Ninety (96%) of these early sites occurred in areas in which the gradient was weak; however, four (4%) occurred in high gradient areas (mean gradient, 71±40 V/cm) at the perimeter of a shock electrode after L:V defibrillations only. Mean gradient at the site of origin of epicardial activity after all successful shocks followed by ectopic activity was 15.4±15.4 V/cm (12.9±6.9 V/cm if the four high gradient sites are omitted).

Eleven episodes of nonsustained, monomorphic ventricular tachycardia occurred after defibrillating shocks. Eight of these episodes occurred after a pause of 167–852 msec after one to seven postshock, ectopic activations originating in low gradient areas (Figure 6). One episode occurred approximately 8 seconds after a successful type B shock, with 6.5 seconds of intervening sinus rhythm before the initiation of ventricular tachycardia. The remaining two episodes occurred 81 and 328 msec after type A defibrillating shocks. All episodes of ventricular tachycardia arose from or near the perimeter of ventricular shock electrodes in which gradients were high (Figure 6). Mean gradient at the site of earliest epicardial activation of these ventricular tachycardias was 47.3±15.5 V/cm. Evidence supporting macroreentry was not identified for any of these episodes of ventricular tachycardia.

**Myocardial and Electrode Impedance**

Electrode impedance accounted for approximately one third to one fourth of total impedance for all
electrode configurations. The mean electrode, cardiac, and total impedance at DFT was generally less with (R+L):V compared with the other configurations (Table 3), although the impedance for (R+L):V was indirectly calculated (see “Methods”). The smaller electrode impedance for (R+L):V was presumably due to the larger combined anodal surface area. At DFT, electrode, cardiac, and total impedance were similar between R:V and the R:V component of the sequential shock; however, all components of impedance were significantly higher for the L:V component of the sequential shock than for L:V shocks alone (p < 0.05). Because impedance decreased as shock strength was increased (see “Appendix”), the lower mean impedances for the L:V shocks, compared with the L:V component of the sequential shock, may simply reflect the larger shock strength at DFT for the single L:V shocks. When L:V shocks of similar voltage to the L:V component of the sequential shock were compared, impedances were not significantly different.

**Discussion**

Defibrillation presumably depends on achieving a sufficient transsarcolemmal potential change across all, or most, of the ventricular myocardium. The transsarcolemmal potential change should be related to the extracellular potential gradient or current density.1,20 There is some indirect evidence to support the theory that a minimum potential gradient or current density is required across all or most of the myocardium to achieve defibrillation. The efficacy of different paddle sizes in transchest defibrillation and of two different external shock electrode locations was shown to be proportional to the generated intracardiac potential gradients.2,3 However, the limited number of electrodes used in these studies did not allow determination of the myocardial distribution of gradients or of the possibility of a minimum gradient necessary for defibrillation. Geddes et al21 using a presumed uniform current density field across an excised, perfused, whole heart in vitro, showed that DFTs in terms of current density were similar for three different waveforms of equal duration. From these studies, Geddes et al estimated that a minimum potential gradient of approximately 10 V/cm was necessary for defibrillation with 10-msec rectangular or trapezoidal waveforms, comparable with the waveform used in this study.

Because the heterogeneous and anisotropic properties of the heart preclude uniform changes in potential, determination of the potential field produced across the ventricles by the shock requires recording potential from multiple sites, which was not done in the above studies. Chen et al2 have attempted to directly measure the distribution of potentials generated across most of the ventricles by defibrillation electrodes; however, the fields recorded were from 1–2-V shocks separated by 1 msec given to R:V and L:V.

*Represents field parameter from maximum combined gradient field (see text).

$\text{\textit{FIGURE 5. Bar graph shows distribution of gradients from all electrode configurations measured at sites of earliest postshock activation (shaded area) and from other early sites of epicardial activation (unshaded area) after shocks. Earliest and other early sites were included from all unsuccessful shocks. Group mean} = 9.5 \pm 5.4 \text{ V/cm; range, 1.7–35.2 V/cm.}$
A shows panel note shock, of Evidence shows potential or an identified. Rapid to adjacent is in the base left apex. FIGURE 6. Panel A: Two-channel recording from right ventricular outflow tract (RVOT) and left ventricular apex adjacent to cathode is shown for a successful 284-V shock with the (R+L):V (dual anode–single cathode configuration) shock from a different experimental animal from the one shown in Figures 2 and 3. Recordings show approximately 500 msec of ventricular fibrillation, which is interrupted by the shock. A single repetitive complex occurs after the shock that had its origin in the base of the right ventricle near the RVOT electrode. This is followed by a regular uniform complex tachycardia with a cycle length of approximately 250 msec and with its origin in the left ventricular apex. Although recordings are bipolar after shock, note prominent ST segment elevation at apical recording site, which is suggestive of myocardial injury. Panel B: Map shows potential gradient field for the same shock. Panel C: Map of second activation sequence after defibrillation shock shown in panel A shows origin of ventricular tachycardia from electrode adjacent to apical shock electrode. Isochronal lines are 10 msec apart. Calculated gradient at this site was 57 V/cm (panel B). Evidence of macroentry in epicardial electrodes was not identified. Rapid activation of most of the apex of the heart suggests an origin deep within ventricles such as from endocardium or septum.

determining the potential gradient field at DFT for several different electrode configurations, the present study provides direct evidence that a minimum potential gradient field is necessary over both ventricles for defibrillation. Despite the marked differences in the electrode configurations used and the differences in their corresponding fields, all obtained similar minimum gradients of approximately 6–7 V/cm at their DFTs. This is similar to the minimum required potential gradient estimated by Geddes et al.11 and to that reported by Witkowski et al.5 Despite the marked difference in total current, voltage, and energy required for defibrillation by L:V compared with the other configurations, the minimum gradients were not significantly different. Also, the efficacy of R:V→L:V appears to be related to its ability to achieve this minimum gradient over both ventricles during at least one component of the sequential shock, although the minimum gradient of either component shock may be considerably less. Thus, it appears that a minimum potential gradient is necessary across all of the ventricular myocardium to achieve defibrillation. Unfavorable electrode configurations require large expenditures of energy or current to obtain this minimum gradient at sites in which their field strength is weak.

The fields generated by electrodes with small surface areas relative to that of the heart are inhomogeneous, as illustrated by the fields generated by all of the electrode configurations used in this study. Because most of the potential is lost near the shock electrodes, high potentials (and thus high energy or current) must be delivered to generate gradients distant to the shock electrodes that are sufficient for defibrillation. Electrodes with larger surface areas, such as those used clinically, should improve the distribution of gradients during shocks and improve defibrillation parameters.2 Small electrodes were used in this study to generate uneven gradient fields to enhance discrimination of defibrillation field requirements and to allow sufficient distribution of epicardial recording electrodes for precise field and activation mapping. The finding that a minimum gradient of 6–7 V/cm is needed for defibrillation should also hold true for larger electrodes.

The DFT was not improved by coupling two anodes together with (R+L):V shocks. Chang et al.12 reported similar results with the use of an intracardiac catheter and either an epicardial or subcutaneous patch electrode. In the present study, the potential gradient field of the (R+L):V shocks was weak at the base of the anterior right ventricle, as was true for the fields of the R:V and L:V shocks. Thus, this particular combination of electrodes did not improve the distribution of gradients across the ventricles. It is also possible that during

<p>| Table 3. Electrode, Cardiac, and Total Impedance at Defibrillation Threshold |
|---------------------------------|-------------------------------|-------------------------------|</p>
<table>
<thead>
<tr>
<th>Anode:Cathode</th>
<th>Electrode (Ω)</th>
<th>Cardiac (Ω)</th>
<th>Total (Ω)</th>
</tr>
</thead>
<tbody>
<tr>
<td>R:V</td>
<td>34.4±9.3</td>
<td>88.8±23.9</td>
<td>123.3±28.0</td>
</tr>
<tr>
<td>L:V</td>
<td>28.4±10.1</td>
<td>79.7±12.2</td>
<td>108.1±18.5</td>
</tr>
<tr>
<td>(R+L):V</td>
<td>27.3±7.0</td>
<td>71.9±15.0</td>
<td>99.2±19.6</td>
</tr>
<tr>
<td>R:V→L:V</td>
<td>32.9±7.7</td>
<td>89.5±20.2</td>
<td>122.4±24.0</td>
</tr>
<tr>
<td>L:V</td>
<td>37.5±11.4</td>
<td>93.4±16.6</td>
<td>130.8±22.4</td>
</tr>
</tbody>
</table>

R:V and L:V, single anode–single cathode configurations; (R+L):V, dual anode–single cathode configuration; R:V→L:V, sequential 7-msec shocks separated by 1 msec given to R:V and L:V.
(R+L):V shocks, vector components of current flow between the R and V electrodes and the L and V electrodes were oppositely directed, particularly between the two anodes in the area of the atrioventricular groove, and that this may have resulted in vector cancellation of these components.

Sequential shocks with short shock separation times to two electrode configurations using an intravascular catheter and patch–electrode configuration or a three-patch–electrode configuration have been shown to have greater efficacy than shocks to single pairs of electrodes.\(^9\) However, not all sequential shock electrode configurations significantly alter the energy requirements for defibrillation.\(^22\) The present study confirmed the greater efficacy of sequential shocks to two different epicardial electrode configurations. The mechanism by which sequential shocks may lower the DFT is presumably due to improvement in the distribution of potential gradients across the ventricles. The L:V component of the sequential shocks at the DFT significantly decreased (by 52%) the number of recording electrodes that measured a gradient less than or equal to 10 V/cm during the preceding R:V component of the sequential shock. Thus, the overall surface area of the ventricles that was exposed to a weak gradient field during the sequential shocks was substantially decreased compared with the R:V shock alone. There was no evidence that the initial R:V shock changed impedance to the subsequent L:V shock alone. However, the minimum gradients of the composite gradient field of the sequential shocks were greater than the minimum gradients of each component shock but were similar to the minimum gradients produced by all of the other single-shock electrode configurations at the DFT. This again suggests the necessity of obtaining a certain minimum gradient over all of the heart for defibrillation. Greater increases in defibrillation efficacy may be obtained by designing electrode configurations in which one shock field is strong where the other is weak, and neither is weak in the same area.

None of the electrode configurations examined in this study produced adequate fields along the right ventricular base, especially along the right ventricular outflow tract. One explanation for this finding is that the three electrodes describe a plane that is located posteriorly (Figure 1B), so that the right ventricular outflow tract, which bulges anteriorly, lies the greatest distance from this plane to produce an area in which the field is weak. This effect may not be as pronounced in the human heart because the right ventricular outflow tract does not project as far anteriorly relative to this plane.

The present study confirms and expands on the finding that ventricular fibrillation first appears after unsuccessful shocks in areas in which the field is weak, regardless of the electrode configuration used.\(^5\) Eighty-seven percent of gradients at the sites of early epicardial activation after unsuccessful shocks were less than 15 V/cm, and 52% were within 1 V/cm of the minimum measured gradient. Sites of initiation of ventricular fibrillation in high gradient areas after unsuccessful shocks were not seen. The lack of a closer correlation between the absolute minimum potential gradient generated by a field and the gradient at the site of resumption of ventricular fibrillation may be explained by a number of factors, such as measurement error and the lack of intramyocardial and endocardial recording sites.

Another factor explaining why earliest activation after an unsuccessful shock is not always in the region exposed to the weakest potential gradient is variation in the electrophysiological state of the myocardium at the time of the shock. Besides the spatial distribution of potential gradients and the temporal waveform of the shock, major determinants of the effect of a shock are probably the state of myocardial activation, refactoriness, and wavefront propagation at the time of the shock. Such factors may prevent the postshock appearance of ventricular fibrillation activation fronts at many sites whether or not the potential gradient of the shock exceeds a minimum threshold. For example, when the region in which the shock field is weakest is in its absolute refractory period at the time of the shock, fibrillation will not reappear from this region even when the potential gradient is below threshold. At another point in time during fibrillation, this region may not be absolutely refractory, and fibrillation activation fronts may originate from this region after the shock,\(^19\) assuming that adjacent sites are capable of conducting the impulse and that surrounding wavefronts do not collide and terminate the reentrant circuit. Consideration of these interactions may help to explain the observation that there exists a range of voltages or energies near threshold in which defibrillation is best described in terms of a probability function.\(^28\)\(^29\) The occurrence of a shock strength with which defibrillation is always achieved and an upper limit of shock strength that will not induce fibrillation no matter when it is introduced into the vulnerable period\(^27\)\(^30\) suggests that there should exist a potential gradient above which fibrillation will be terminated regardless of the state of the myocardium. However, because the shock strength that is always successful is usually greater than that at the DFT,\(^28\)\(^29\) the minimum gradient measured in this study is probably less than that required to always defibrillate.

The occurrence of a minimum gradient at the DFT is most consistent with either the total ventricular depolarization (or extinction) hypothesis of defibrillation\(^22\) or the upper limit of vulnerability hypothesis,\(^19\)\(^27\)\(^30\) because both theories predict a minimum required potential gradient across all of the ventricular myocardium. The finding of a required minimum gradient across all of the ventricular myocardium is less consistent with the critical mass hypothesis.\(^19\)\(^33\) The latter hypothesis predicts that the minimum necessary gradient for different electrode configurations should be equal, not over the entire ventricular myocardium but at the boundary enclosing the critical mass. The absolute minimum gradient would not necessarily have to be equal for different electrode configurations, because it would depend only on the rate of decline in the gradient field beyond the border of the critical mass. The critical mass for defibrillation has been estimated to be approximately 75% of the ventricular myocardium with prolonged depolarization with potassium administration.\(^33\)

If so, equal potential gradients should be found at the border of the regions enclosing 75% of the ventricular myocardium for different electrode configurations, not at the minimum gradient value over all of the ventricles.
The findings in this study that 1) the L:V shock had a smaller surface area of low gradients compared with the other shocks (Figure 4), despite the fact that it was markedly less efficient in defibrillation, and 2) the minimum potential gradients are not significantly different for the different lead configurations both run contrary to the predictions of the critical mass hypothesis. However, because endocardial and septal gradient measures are not available for determination of the three-dimensional potential gradient field, it is possible that intramyocardial potential gradient distributions do not parallel in a relatively symmetrical manner the distribution measured on the epicardium and that a critical mass less than 100% existed but was not identified in this study. We hope that further developments in the calculation of gradients across the entire volume of the heart will help to resolve this issue.34

A potential gradient greater than approximately 5 V/cm for a 3-msec, low-tilt, monophasic shock delivered during the relative refractory period of a passing wavefront induces conduction block, presumably by prolonging refractoriness; gradients less than this do not produce conduction block.35 This critical interaction between refractoriness and decreasing field strength may lead to the induction of reentry and ventricular fibrillation. According to the upper limit of vulnerability hypothesis, subthreshold shocks will halt all fibrillation wavefronts but reinitiate ventricular fibrillation, presumably by a mechanism similar to that just described.19 The similarity between the minimum gradient of 6–7 V/cm for defibrillation found in this study and the critical gradient of 5 V/cm for electrical induction of fibrillation in the vulnerable period of regular rhythm is consistent with but does not prove this hypothesis.

There are several limitations to this study. Waveforms of depolarization are obscured during defibrillation by the shock and immediately after the shock by the baseline deflection generated by switching off the mapping system modification; thus, depolarization during or immediately after the shock cannot be identified or mapped. Furthermore, the lack of septal, intramyocardial, and endocardial recording sites limits the resolution of the mapping and the determination of all components of the gradients. Specifically, the lack of transmural recordings may decrease the absolute value of the measured gradients. For low-voltage shocks to an electrode array resembling R:V, transmural gradients in the right ventricular outflow tract are approximately half the tangential gradient.4 The exposure of the anterior surface of the heart to air in the open-chest dog preparation may increase the anterior gradient measurements due to a boundary effect; however, the greater tangential current flow would improve gradient calculation using epicardial recording electrodes. Measurement of rapidly changing gradients near shock electrodes requires small interelectrode distances to maintain accuracy; irregularities in high gradient areas in this study may reflect inadequate spacing of recording electrodes or inhomogeneous current distribution around the perimeter of the shock electrode. Depolarization of ventricular myocardium is dependent on fiber orientation relative to the vector of the potential gradient.36 Whether fiber orientation affects the field requirements for defibrillation or whether different areas of the ventricles have the same field requirements were not assessed in this study.

When only a certain minimum gradient is needed for defibrillation, electrode configurations that must generate very high gradients near the shock electrodes to achieve this minimum gradient distantly not only waste energy but increase the risk of inducing variable degrees of myocardial injury. High potential gradients may cause cell injury and death, decreased contractility, asystole, conduction block, and dysrhythmias.1,37–42 Extremely high gradients may even generate intractable ventricular fibrillation secondary to induced injury.30,40 Despite the great unevenness of the fields generated by the electrode configurations used in this study, postshock arrhythmias occurred relatively infrequently. There was no evidence that the high gradients at the shock electrodes directly initiated ventricular fibrillation after unsuccessful shocks, although repetitive activity occurred infrequently in high gradient areas after successful shocks. Specifically, this occurred only after successful L:V shocks, which generated the highest gradients for defibrillation, perhaps reaching a specific injury threshold. High-voltage shock induction of ventricular fibrillation (type II ventricular fibrillation) probably requires voltages higher than those used in this study.30,40 Although shock-induced arrhythmias may occur clinically41 and have been shown to occur more frequently after high-voltage shocks in vitro,1,37,40 the present study is the first to map the origin of postshock ventricular tachycardia from the site of the shock electrode, which in all cases was recorded from electrodes at the perimeter of the shock electrode possibly reflecting the higher current density at the electrode edge.42 The mechanism of the postshock ventricular tachycardia seen in this study could not be ascertained; evidence of epicardial macroreentry was not seen.

The ability to measure shock potentials also allows measurement of various components of impedance, such as that at the electrode–myocardial interface or across the heart itself (see “Appendix”). The impedance measured in this study is generally greater than that reported by others for transmyocardial shocks,44,45 presumably reflecting the small size of the shock electrodes used in this study. Unlike a previous study,27 the present study did not show significant changes in impedance with increasing shock voltage for R:V, (R+L):V, or the components of R:V→L:V. This may be due to the random sequence of shock delivery and/or to the relatively small range of shock voltages delivered to these configurations. There was a significant change with L:V, which, because of its decreased efficacy, required a larger number of shocks over a greater range of voltages to achieve a DFT. When R:V shocks were given over a greater range of voltages, marked changes in impedance were noted (see “Appendix”). However, Lawrence et al44 were not able to show a change in impedance between shocks of different strengths when more than 5 minutes elapsed between shocks and showed only a very small change when the shocks were separated by a few seconds.

The ability to map the potential gradient field generated by shock electrodes should lead to marked improvements in their design and function. If the desired goal is a uniform field slightly exceeding a certain minimum gradient, the possible electrode combinations
are almost limitless; however, application of the rule of superposition may lead to a more efficient way of testing potential fields for numerous electrode configurations with a relatively limited number of shocks. Further description of factors involved in electrode impedance may also result in marked improvement in electrode design. If electrodes, electrode configurations, and shock sequences can be designed that will generate a more even field that only slightly exceeds the minimum gradient required for reliable defibrillation without increasing electrode impedance, implantable defibrillators then can be designed that will require less energy to operate. This in turn will have major benefits, such as increasing the lifetime and decreasing the size of the battery of implanted units and decreasing the pain, potential cardiac dysfunction, and dysrhythmias associated with defibrillation shocks.

Appendix
Effect of Shock Strength on Impedance
In an earlier study with low-voltage shocks from 1 to 2 V given to R: V using 4.5-cm² titanium mesh electrodes, only 17.1±3% of the delivered shock potential was measured on the ventricles, whereas in the present study, recorded myocardial potential was 76.4±6.3% of that delivered with 2.64-cm² electrodes. The addition of recording electrodes beneath the shock electrodes and inclusions of the voltage drop across the atria in this study may explain some but not all of this difference. To see if part of this discrepancy was caused by differences in impedance for low- and high-voltage shocks, an additional dog was studied with the same surgical preparation as described in “Methods.” However, 4.5-cm²-round titanium mesh shock electrodes with a single button electrode attached to the ventricles with a 2.64-cm² round titanium mesh electrode underneath were attached in an R: V configuration. Shocks of 14-msec duration were delivered once each at increasing strengths as follows: 1, 2, 3, 4, 5, 6, 7, 8, 9, 10, 12.5, 15, 20, 25, 50, 75, 100, 150, 200, 300, 400, 500, 750, and 1,000 V. Shocks from 1 to 25 V were delivered by a constant voltage device and from 25 to 1,000 V by the device used in the present study. Two calibrated input dynamic ranges (±10 V and ±500 V) to the amplifiers were used to record potentials. Current was measured on an oscilloscope.

With 1–4-V shocks, total impedance increased markedly from 254 to 434 Ω, decreased abruptly to 273 Ω at 5 V, and then gradually declined with increasing potential down to 106 Ω at 1,000 V. A plot of the logarithm of potential delivered versus the calculated total impedance illustrates the marked change in impedance at low voltages (Figure 7). Beyond approximately 10 V, the relation between impedance and the logarithm of delivered voltage became more linear.

To define where the change in impedance occurred, mean differences in potential between successive rows of electrodes starting at the apical shock electrode were calculated and divided by the total measured current as an estimate of effective impedance of the layers of ventricular myocardium between each row of electrodes. The greatest impedance was calculated between the shock electrode and the recording electrode underneath it (row 1) for shocks of 1–20 V (Figure 7). At 25 V, however, the impedance at row 1 and the next most cephalad row (row 2) equalized with further increases in delivered potential causing the greatest impedance change at row 2, principally because impedance at row 1 continued to decrease, whereas it was relatively stable at row 2. The marked increase in impedance at 4 V is reflected in the impedance curves at all levels of the ventricles, although most of this occurs in the apical third of the myocardium in rows 1–3 (Figure 7).

For a 1-V R: V shock, 36% and 15% of delivered potential was measured between electrodes beneath (row 1) and immediately encircling (row 2) the apical shock electrode, respectively, and the electrode beneath the right atrial anode. These measurements were not significantly changed at 4 V and were in accordance with the previous study. As delivered potential increased from 5 V, 25 V, 100 V, and 1,000 V, the percent of delivered potential measured in row 1 steadily increased to 48%, 66%, 73%, and 75% and in row 2 from 20%, 32%, 34%, and 43%, respectively.

The current waveform for shocks less than 5 V showed a logarithmic increase in current flow during the period of the shock, suggesting a large contribution of capacitance to the overall impedance. This capacitive element may explain the initial increase in total impedance. As potential was increased above 5 V, the current waveform showed progressively less distortion and impedance decreased, primarily at or near the electrode–myocardial interface. This strong capacitive element to impedance may be generated to some extent by electrode polarization at the myocardial interface during low-voltage shocks. However, the occurrence of similar changes in estimated impedance distant to the apical shock electrode implies myocardial changes occurring as a function of increasing potential delivered. This capacitive component of impedance progressively decreased with increasing delivered potential so that in the range of potentials necessary for defibrillation with epicardial electrodes, impedance appears to be primarily resistive, confirming the work of others.

Acknowledgments
We would like to thank Joseph C. Greenfield Jr. for his support and advice; and Ellen G. Dixon, Sharon B. Melnick,
Yohannes Afework, Alton T. Ledford, and Cloyce M. Lassiter for their technical assistance.

References

26. Mehra R, Maracaccini S: Comparison of sequential and simultaneous pulse defibrillation threshold with a non-epicardial electrode system (abstract). Circulation 1986;74(suppl II):11-184
32. Wijgers CJ: The physiologic basis for cardiac resuscitation from ventricular fibrillation: Method for serial defibrillation. Am Heart J 1940;20:413–422
Cardiac potential and potential gradient fields generated by single, combined, and sequential shocks during ventricular defibrillation.

J M Wharton, P D Wolf, W M Smith, P S Chen, D W Frazier, S Yabe, N Danielely and R E Ideker

*Circulation*. 1992;85:1510-1523
doi: 10.1161/01.CIR.85.4.1510

*Circulation* is published by the American Heart Association, 7272 Greenville Avenue, Dallas, TX 75231

Copyright © 1992 American Heart Association, Inc. All rights reserved.

Print ISSN: 0009-7322. Online ISSN: 1524-4539

The online version of this article, along with updated information and services, is located on the World Wide Web at:

http://circ.ahajournals.org/content/85/4/1510

Permissions: Requests for permissions to reproduce figures, tables, or portions of articles originally published in *Circulation* can be obtained via RightsLink, a service of the Copyright Clearance Center, not the Editorial Office. Once the online version of the published article for which permission is being requested is located, click Request Permissions in the middle column of the Web page under Services. Further information about this process is available in the Permissions and Rights Question and Answer document.

Reprints: Information about reprints can be found online at:

http://www.lww.com/reprints

Subscriptions: Information about subscribing to *Circulation* is online at:

http://circ.ahajournals.org//subscriptions/