Some factors affecting bubble formation with catheter-mediated defibrillator pulses

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ABSTRACT  Factors affecting bubble formation during delivery of defibrillator pulses to arrhythmogenic cardiac tissue via a catheter are unknown. We investigated the role of energy, electrode surface area, interelectrode distance, and electrode polarity on bubble formation and on current and voltage waveforms during delivery of damped sinusoidal discharges from a standard defibrillator to anticoagulated bovine blood. Gas composition was studied with mass spectrometry. Defibrillator energy settings were varied between 5 and 360 J. The principal catheter used for study was a Medtronic 6992A lead. Additional electrodes tested included 2, 5, and 10 mm long No. 6F, 7F, and 8F copper electrodes. Interelectrode distances used to assess the effect of anode-cathode spacing were 1, 5, 10, and 20 cm. Bubble volume increased linearly from 0.043 to 0.134 ml per cathodal pulse and from 0.030 to 3.50 ml per anodal pulse as energy settings were increased from 5 to 360 J (r = .99). Typical smooth waveforms for both current and voltage were seen only in the absence of bubbles. The voltage waveform was distorted for each cathodal pulse of 100 J or more and for each anodal pulse of 10 J or more only if bubbles were present. The effect of electrode surface area on bubble formation was tested at a 200 J energy setting and at a 10 cm interelectrode distance with the use of cathodal pulses. Bubble formation varied inversely with electrode surface area (r = .876). Bubble formation, however, varied minimally as interelectrode spacing was changed from 1 to 20 cm. The effect of polarity on bubble formation when the Medtronic 6992A distal electrode and an 8.5 cm disk electrode separated by 10 cm were used was highly significant. For a 200 J pulse, bubble formation with the catheter as anode was 3.30 ± 0.10 ml and with the catheter as cathode it was 0.070 ± 0.002 ml (p < .001). Mass spectrometry of both anodal and cathodal gas samples demonstrated the constituents of the gas bubble to include a variety of gases, which is inconsistent with simple electrolytic production of the bubbles observed. The predominance of nitrogen in either polarity sample suggested that the principal source of the bubble was dissolved air. In summary, bubble formation at an electrode receiving damped sinusoidal outputs from a standard defibrillator does not vary significantly with varying interelectrode distance. However, it is directly proportional to energy and inversely proportional to electrode surface area. Anodal catheter discharges produce considerably more bubbles than do cathodal discharges. Distortions in the voltage waveform correlate with physical factors leading to high-pressure shockwave generation and to subsequent extrusion of dissolved gases from solution.


GAS EMBOLIZATION is a potential concern after delivery of catheter-mediated defibrillator pulses to cardiac tissue.1–3 This issue is most relevant when pulses are delivered in the left ventricular cavity for ablation of ventricular tachycardia foci.4–6 Bubble formation in this chamber might embolize to the coronary arteries or to the systemic circulation, raising questions regarding the safety of the present technique, especially in view of previous work in dogs that has suggested that 0.1 to 0.2 ml of air embolized into the coronary artery is sufficient to result in myocardial damage.9

To determine whether potentially hazardous gas volumes could be generated in the heart during catheter-mediated defibrillator pulses, we investigated several factors in vitro that might influence gas formation. The factors investigated included: defibrillator energy, electrode size or surface area, interelectrode spacing,
and polarity. The effects of gas formation on current and voltage waveforms also were observed. Mass spectroscopic findings of the bubble gas together with waveform analysis, pressure recordings, and review of previous work in plasma physics and electric pulse technology provided insight into the mechanisms of bubble genesis with this technique.

**Methods**

Experiments were performed in a 26 × 30 × 50 cm tank filled with fresh, anticoagulated bovine blood. Damped sine-wave pulses (Edmark waveform) were delivered between electrodes with a standard defibrillator (Physio-Control Life Pak 6) as the energy source. Proper defibrillator function was verified before experimentation by testing its output into a 50 Ω resistor. Bubbles from each pulse were collected in a submerged wide-mouth, inverted funnel with a capped, calibrated stem centered above the electrode (figure 1). Any gas that was generated dissolved in the medium as it rose into the cylindrical neck of the funnel, allowing for gas volume measurement and collection of the gas for subsequent analysis. A surfactant was used to prevent foaming of the blood during the study. The basic experiments studying the effect of energy, surface area, interelectrode distance, and polarity on bubble generation were performed in a room air atmosphere in bovine blood. However, an argon rather than room air atmosphere, and saline rather than bovine blood, were used in additional polarity experiments in order to better understand the determinants of gas generation. To produce the argon atmosphere, the tank was surrounded by a plastic enclosure that was filled with pure argon for 30 min.

The gas composition was determined with a Balyers QMG 511 quadrupole mass spectrometer. Quantitative analysis of the gas components was possible by comparison of the mass spectrometry signal from the study gas against that from room air samples for which gas concentration was known. Previous mass spectroscopy work relating to the analysis of hydrogen, nitrogen, and oxygen was also used in determining composition of the gas generated. The Medtronic 6992A catheter provided the principal electrode used for the study. This catheter was selected because of its high dielectric strength (over 5000 V) and its ability to withstand multiple high-energy pulses without changes in line resistance, alterations in its insulation, or openings at its connecting cable-electrode weld. Pulses were delivered to the Medtronic 6992A electrodes via both the connector pin and the stylet. By use of the stylet, in addition to the connector pin, lead resistance diminished and resulted in resistance values during pulse delivery comparable to those previously observed with standard pacing and recording wires. Standard pacing and recording catheters were not used in this study because previous investigations demonstrated their inadequate dielectric properties and their propensity for electrode breakdown when subjected to defibrillator pulses. Because a variety of suitable clinical electrodes were not available for those studies investigating the effect of electrode size (surface area) on bubble formation, copper wire electrodes with gauges similar to No. 6F, 7F, and 8F catheters were used in addition to the Medtronic catheter. Surface area was varied with these copper electrodes by adjusting the length of uninsulated wire exposed at the tip for each of the 6F, 7F, and 8F cables used. Electrode lengths from the tip of the cable to insulation were 2, 5, and 10 mm. The 2 mm length was chosen to represent the length of a standard catheter-tip electrode. The 5 and 10 mm lengths were chosen as means to expand surface area without the use of impractically large electrodes. A stainless steel disk 8.5 cm in diameter was used to represent the chest electrode.

During each pulse, the current waveform was displayed on a Tektronix oscilloscope (model No. 5111A) with the use of a Tektronix current probe (model No. A6303). Voltage was displayed simultaneously with the current on the oscilloscope with the use of a resistive voltage divider with a 1000:1 input-to-output ratio (figure 1). Peak voltage and peak current were measured for each pulse from the oscilloscope screen. The contour of the current and voltage waveforms also were studied and compared with the normally smooth damped sinusoidal waveform usually seen from the Physio-Control Life Pak 6 when discharged into a 50 Ω resistor. Delivered energy for each pulse was determined by integrating the product of the

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**FIGURE 1.** Experimental schema for collection of bubbles and analysis of voltage and current waveform during delivery of defibrillator pulses to catheters (see methods section). The tank was filled with either saline or bovine blood.

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current and voltage waveform over the duration of the pulse (Physio Control, Waveform Integrator). Pressure changes as a consequence of the pulse were measured with a Piezotronics piezoelectric pressure transducer.

The effects of variations in (1) energy, (2) electrode size or surface area, (3) interelectrode distance, and (4) polarity on gas formation were studied as follows.

(1) In the first experiment, gas formation was evaluated as a function of energy and polarity with the Medtronic 6992A distal electrode. The Medtronic 6992A lead has two electrodes. The distal electrode used in this part of the study has a surface area of 19 mm². The Medtronic catheter electrode was used initially in the cathodal(−) configuration and the 8.5 cm disk was used in the anodal(+) configuration. The anode-cathode interelectrode distance was fixed at 10 cm. Energy settings tested were 5, 10, 20, 50, 100, 200, 300, and 360 J. Subsequently, polarity was switched and the Medtronic 6992A distal electrode served as the anode and the experiment was repeated over the energy settings listed above.

(2) Gas formation as a function of electrode size or surface area was assessed in the cathodal configuration for each of the available electrodes at a fixed energy setting of 200 J. The anode was the 8.5 cm disk separated from the cathode by a fixed distance of 10 cm. The catheter was cathodal in this experiment because of the findings observed in the polarity studies. The following electrodes were used as cathodes: Medtronic 6992A distal electrode, Medtronic 6992A proximal electrode, combined distal and proximal Medtronic 6992A electrodes connected in parallel, No. 6F, 7F, and 8F copper electrodes, each having electrode lengths of 2, 5, and 10 mm.

(3) Gas formation as a function of interelectrode distance was assessed at 1, 5, 10, and 20 cm separation at a fixed energy setting of 200 J with the distal pole of the Medtronic 6992A lead as the cathode and the 8.5 cm disk as the anode.

(4) The effect of polarity on gas formation was assessed both in saline and blood with the use of the distal pole of the Medtronic 6992A lead and the 8.5 cm disk. Each electrode was used alternately as either the anode or the cathode. In addition, gas formation was assessed with two 6992A catheters in which the distal electrode of one served as the anode and the distal electrode of the other served as the cathode. The interelectrode distance was 10 cm. The defibrillator energy setting was constant at 200 J. Saline, as well as blood, was used for this study to further define the mechanism of gas formation and test saline’s reliability as a substitute for blood in the study of physical mechanisms involved with catheter-mediated electrical pulses. For one of the polarity studies, the Medtronic 6992A lead was substituted for a No. 6F, 2 mm copper electrode. Polarity experiments also were performed in an argon atmosphere with the use of the same electrode configurations, defibrillator energy settings, and interelectrode distance as described above to better determine the source of the gas components.

Because gas volume could be small, especially with smaller energy pulses, and therefore difficult to quantify for a single pulse, gas was collected over 10 to 30 pulses for each test and then averaged per pulse. Linear regression techniques were used to correlate the effect of delivered energy, electrode surface area, and interelectrode distance on gas volume. A Student t test was used to compare the differences between anodal and cathodal pulses for the polarity studies.

Results

The results from the studies correlating gas formation as a function of energy are shown in Table 1 for both cathodal and anodal catheter discharges. With either polarity, the gas volume per pulse rose as the delivered energy level rose. However, it was apparent that a distinct level of delivered energy needed to be reached before any measurable amount of gas was generated. Cathodal pulses required a higher delivered energy than anodal pulses (78 vs 12 J) (p < .001) to generate gas. Once gas formation occurred, the bubble volume per pulse could be predicted for any given delivered energy for each electrode polarity used. With

<table>
<thead>
<tr>
<th>TABLE 1</th>
<th>Gas formation in blood medium as a function of energy</th>
</tr>
</thead>
<tbody>
<tr>
<td>Anode (+)</td>
<td>Cathode (−)</td>
</tr>
<tr>
<td>8.5 cm disk</td>
<td>Distal 6992A</td>
</tr>
<tr>
<td>8.5 cm disk</td>
<td>Distal 6992A</td>
</tr>
<tr>
<td>8.5 cm disk</td>
<td>Distal 6992A</td>
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<tr>
<td>Distal 6992A</td>
<td>8.5 cm disk</td>
</tr>
<tr>
<td>Distal 6992A</td>
<td>8.5 cm disk</td>
</tr>
</tbody>
</table>

TSTM = too small to measure.

Interelectrode distance: 10 cm.
the 6992A distal electrode as the cathode and the 8.5 cm disk as the anode, bubble volume was predicted by the following regression equation:

Bubble volume/pulse (ml) = 0.006 + (0.0004 delivered energy in J)
\[ r = 0.993, p < .001 \]

For the configuration in which the 6992A distal electrode served as the anode and the 8.5 cm disc served as the cathode, the regression equation for bubble volume was:

Bubble volume/pulse (ml) = −0.22 + (0.019 delivered energy in J)
\[ r = 0.999, p < .001 \]

The addition of other predictor variables — peak voltage, peak current, energy setting — to either model did not improve the prediction of bubble volume, but as single predictors these were nearly as good as energy delivered.

For cathodal discharges, gas volume per joule was relatively constant (approximately \(5 \times 10^{-4}\) ml/J). However, with anodal discharges, gas volume per joule rose in a nonlinear fashion to values as large as \((187 \pm 12) \times 10^{-4}\) ml/J. Anodal pulses above 200 J settings generated such powerful shock waves that the collection apparatus was shattered, preventing collection of the gas volume generated. Examples of shock waves recorded with some of the higher energy pulses are shown in figure 2. Panel A is a recording of a pressure pulse 2 cm from a cathodal electrode receiving a 200 J stored pulse. The peak pressure for this cathodal discharge is 18.9 atmospheres or 14,364 mm Hg. Panel B shows the pressure from a 200 J setting anodal pulse at 2 cm from the electrode. For the anodal discharge, pressure rose to 38.5 atmospheres or 29,260 mm Hg.

For either polarity, all discharges associated with significant gas formation had a distortion in the voltage waveform characterized first by a rise in voltage over a brief time period (figure 3). The current waveform, because of the inertia of the defibrillator inductor, remained relatively faithful to the expected Edmark configuration. The additional rise in voltage without a concomitant rise in current indicates an abrupt rise in impedance during the pulse. To illustrate the relationship between waveform configuration and delivered energy, arcing, and gas formation, figure 3 contains examples of the arc, bubbles, and waveforms for defibrillator settings of 5 to 360 J. For demonstration purposes, these photographs were taken in saline to visualize the arcing and bubble formation that otherwise would not have been possible in the opaque medium of blood. Note the differences between gas volume generated in saline (as shown in figure 3) as compared with that generated in blood (table 1). Speculations as to why saline and blood yield such disparate results are outlined in the discussion section.

For any particular energy setting on the Physio-Control defibrillator dial, delivered energy could vary considerably from expected values, especially with the higher energy cathodal discharges (table 1). However, once delivered energies were measured for the particular set of circumstances chosen for this experiment, a regression equation could be generated that allowed delivered energy to be predicted from the energy setting for either polarity. For the situation in which the
FIGURE 3. Flash, bubbles, and voltage and current waveforms as a function of increasing energy settings for defibrillator pulses delivered with a Life-Pak 6 Physio-Control defibrillator to a No. 6F, 2 mm long Cu anodal electrode. The cathode is an 8.5 cm diameter disk separated from the anode by 10 cm. The actual results from blood experiments investigating the effect of energy on gas generation are listed in table 1. However, because blood is opaque, for illustrative purposes we present photos taken from a similar study performed in saline to demonstrate the events at the time of electric discharge. Note that the amount of gas generated is considerably less than that observed in blood as presented in table 1 and emphasizes the failure of experiments in saline to predict outcomes in blood reliably. Nevertheless, the processes taking place are similar. The flashes are a result of arcing (see discussion for explanation) and are shown on the left, the resultant bubbles seen within 1 sec later are shown in the middle, and the associated voltage and current waveforms are shown on the right. The numbers shown above the voltage and current waveforms are the energy setting, peak delivered current, peak delivered voltage, delivered energy, and bubble volume. A through F demonstrate the effects of 5, 20, 50, 100, 200, and 360 J pulses, respectively. A, A 5 J pulse does not result in arcing or gas formation, and the voltage and current waveforms maintain their typical damped sinusoidal configurations. B through F, Arcing and gas formation are observed. Note that gas formation is evident within the center of the flash and is especially visible during the 360 J discharge in F. (The gas is the light inner image within the flash that appears to rise up.) The degree of arcing and gas formation increases as energy increases. The voltage waveform shows a rise or "peak" two-thirds to one-third of the way through the discharge, which probably correlates to a rise in impedance from electrolysis gas insulating the electrode (see text for discussion). After approximately 4 msec into the waveform, there is an abrupt cessation of current flow. This is automatically determined by the defibrillator and a mechanical relay switch that can vary in response from defibrillator to defibrillator as well as with current waveform. After automatic discontinuation of the pulse by the defibrillator there is some bounce of the relay switch and current again flows briefly.

6992A electrode serves as the cathode, the regression equation predicting delivered energy was:

\[ \text{Delivered energy (J)} = 3.04 + (0.76 \text{ energy setting}) \]

\[ r = 0.99, p < 0.001 \]

For the situation in which the 6992A electrode served as the anode, the regression equation predicting delivered energy was:

\[ \text{Delivered energy (J)} = 3.61 + (0.92 \text{ energy setting}) \]

\[ r = 0.99, p < 0.001 \]

Polarity also affected voltage and current. For any given delivered energy, peak delivered voltage was higher and peak delivered current was lower for anodal pulses than for cathodal pulses. These findings indicate that higher resistance to current flow occurs with anodal discharges.

In table 2, bubble formation is displayed as a function of the electrode surface area. Although the copper wire and the Medtronic catheters are not directly com-
TABLE 2

Gas formation as a function of electrode surface area

<table>
<thead>
<tr>
<th>Cathode (−)</th>
<th>Cathode surface area (mm²)</th>
<th>Inter-electrode distance (cm)</th>
<th>Energy setting (J)</th>
<th>Delivered energy (J)</th>
<th>Peak voltage (V)</th>
<th>Peak current (A)</th>
<th>Bubble volume/pulse (ml)</th>
<th>Bubble volume/J (ml/J)</th>
<th>Voltage waveform distortion</th>
</tr>
</thead>
<tbody>
<tr>
<td>6F, 2 mm Cu wire</td>
<td>12</td>
<td>10</td>
<td>200</td>
<td>140 ± 5</td>
<td>1112 ± 35</td>
<td>74 ± 1</td>
<td>0.076 ± 0.002</td>
<td>(5.4 ± 0.3) × 10⁻⁴</td>
<td>Yes</td>
</tr>
<tr>
<td>6F, 5 mm Cu wire</td>
<td>28</td>
<td>10</td>
<td>200</td>
<td>138 ± 6</td>
<td>1013 ± 26</td>
<td>75 ± 1</td>
<td>0.048 ± 0.002</td>
<td>(3.4 ± 0.3) × 10⁻⁴</td>
<td>Yes</td>
</tr>
<tr>
<td>6F, 10 mm Cu wire</td>
<td>53</td>
<td>10</td>
<td>200</td>
<td>142 ± 9</td>
<td>988 ± 11</td>
<td>74 ± 1</td>
<td>0.037 ± 0.005</td>
<td>(2.6 ± 0.6) × 10⁻⁴</td>
<td>Yes</td>
</tr>
<tr>
<td>7F, 2 mm Cu wire</td>
<td>16</td>
<td>10</td>
<td>200</td>
<td>135 ± 8</td>
<td>1035 ± 26</td>
<td>75 ± 1</td>
<td>0.056 ± 0.002</td>
<td>(4.1 ± 0.4) × 10⁻⁴</td>
<td>Yes</td>
</tr>
<tr>
<td>7F, 5 mm Cu wire</td>
<td>36</td>
<td>10</td>
<td>200</td>
<td>139 ± 6</td>
<td>1004 ± 21</td>
<td>75 ± 1</td>
<td>0.042 ± 0.001</td>
<td>(3.0 ± 0.2) × 10⁻⁴</td>
<td>Yes</td>
</tr>
<tr>
<td>7F, 10 mm Cu wire</td>
<td>68</td>
<td>10</td>
<td>200</td>
<td>141 ± 9</td>
<td>997 ± 16</td>
<td>75 ± 1</td>
<td>0.039 ± 0.001</td>
<td>(2.8 ± 0.2) × 10⁻⁴</td>
<td>Yes</td>
</tr>
<tr>
<td>8F, 2 mm Cu wire</td>
<td>22</td>
<td>10</td>
<td>200</td>
<td>141 ± 5</td>
<td>1052 ± 26</td>
<td>75 ± 1</td>
<td>0.052 ± 0.001</td>
<td>(3.7 ± 0.2) × 10⁻⁴</td>
<td>Yes</td>
</tr>
<tr>
<td>8F, 5 mm Cu wire</td>
<td>46</td>
<td>10</td>
<td>200</td>
<td>140 ± 5</td>
<td>1038 ± 24</td>
<td>74 ± 1</td>
<td>0.041 ± 0.002</td>
<td>(2.9 ± 0.2) × 10⁻⁴</td>
<td>Yes</td>
</tr>
<tr>
<td>8F, 10 mm Cu wire</td>
<td>87</td>
<td>10</td>
<td>200</td>
<td>147 ± 10</td>
<td>1000 ± 27</td>
<td>75 ± 1</td>
<td>0.026 ± 0.001</td>
<td>(1.8 ± 0.2) × 10⁻⁴</td>
<td>Yes</td>
</tr>
<tr>
<td>Distal 6992A</td>
<td>19</td>
<td>10</td>
<td>200</td>
<td>172 ± 2</td>
<td>1557 ± 37</td>
<td>61 ± 1</td>
<td>0.068 ± 0.001</td>
<td>(4.0 ± 0.1) × 10⁻⁴</td>
<td>Yes</td>
</tr>
<tr>
<td>Proximal 6992A</td>
<td>48</td>
<td>10</td>
<td>200</td>
<td>172 ± 6</td>
<td>1503 ± 87</td>
<td>61 ± 1</td>
<td>0.036 ± 0.001</td>
<td>(2.1 ± 0.1) × 10⁻⁴</td>
<td>Yes</td>
</tr>
<tr>
<td>Distal and proximal 6992A</td>
<td>67</td>
<td>10</td>
<td>200</td>
<td>167 ± 6</td>
<td>1451 ± 35</td>
<td>65 ± 1</td>
<td>0.032 ± 0.002</td>
<td>(1.9 ± 0.2) × 10⁻⁴</td>
<td>Yes</td>
</tr>
<tr>
<td>8.5 cm disk</td>
<td>5630</td>
<td>10</td>
<td>200</td>
<td>81 ± 3</td>
<td>450 ± 0</td>
<td>88 ± 0</td>
<td>—</td>
<td>—</td>
<td>No</td>
</tr>
</tbody>
</table>

The anode was an 8.5 cm disk in each case.

parable, the same trend in the volume of gas generated per square millimeter existed for either type of electrode. Bubble volume decreased as the surface area increased, even though delivered energy, peak delivered voltage, and peak delivered current remained relatively constant for each electrode type used. The increase in delivered energy and in peak voltage with the Medtronic lead is probably secondary to higher lead resistances. In general, however, gas volume generated per unit of delivered energy is similar for comparable surface area electrodes (e.g., the No. 7F, 2 mm copper and the 6992A distal Medtronic electrode). For either catheter studied, bubble volume per pulse could be predicted from electrode surface area as follows:

Bubble volume/pulse (ml) = 0.07 − (0.0005 surface area in mm²)

r = .876, p < .001

These findings suggest that energy density is a critical factor in gas formation.

In the example in which an 8.5 cm disk was used as both the anode and the cathode, the large surface area strongly influenced the amount of gas generated. In this case, resistance to current flow fell and there was a marked decrease in the peak voltage and in the amount of delivered energy, with a significant rise in the peak delivered current. These findings imply that with electrodes of smaller surface area, much of the pulse energy is dissipated around the electrode in the process of gas formation and in the arc.

Table 3 demonstrates the relationships for gas formation as a function of interelectrode spacing. The distance between anode and cathode had only a slight effect on gas formation, with increasing interelectrode distances resulting in less gas even though delivered energy rose somewhat. As the volume of blood between anode and cathode increased, there was a lower peak current and a higher peak voltage consistent with higher resistances. For the particular circumstances chosen for study, the regression equation for bubble volume per pulse as a function of interelectrode distance is as follows:

Bubble volume/pulse (ml) = 0.07 − (0.0003 distance in cm)

r = .846, p < .001

Table 4, along with the findings in table 1, shows
The anode was the 8.5 cm disk, the cathode the distal 6992A, and the energy setting 200 J in each case.

The No. 6F, 2 mm copper wire electrode (table 4) demonstrated polarity-dependent findings similar to those seen with the more clinically applicable Medtronic 6992A lead. The differences in gas formation between the copper electrode and the Medtronic electrode were small and most likely related to differences in electrode surface area and lead resistance.

When both the anode and the cathode were catheter electrodes, instead of one being the 8.5 cm disk, bubbles arose from both poles. The amount, however, was slightly less than what would have been observed with the standard catheter-disk configuration, particularly in blood. Catheter-catheter discharges provide higher resistance to current flow, and delivered energy is greater — probably with more energy dissipated in the arc and in the process of bubble formation.

The experiments in an argon atmosphere (table 4) demonstrate similar gas volume findings to those ob-

### TABLE 3
Gas formation in blood medium as a function of interelectrode spacing

<table>
<thead>
<tr>
<th>Inter-electrode distance (cm)</th>
<th>Delivered energy (J)</th>
<th>Peak voltage (V)</th>
<th>Peak current (A)</th>
<th>Bubble volume/pulse (ml)</th>
<th>Bubble volume/J (ml)</th>
<th>Voltage waveform distortion</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>167 ±5</td>
<td>1485 ±19</td>
<td>65 ±1</td>
<td>0.073 ±0.001</td>
<td>(4.4 ±0.2) x 10^{-4}</td>
<td>Yes</td>
</tr>
<tr>
<td>5</td>
<td>168 ±2</td>
<td>1501 ±23</td>
<td>62 ±1</td>
<td>0.072 ±0.001</td>
<td>(4.3 ±0.1) x 10^{-4}</td>
<td>Yes</td>
</tr>
<tr>
<td>10</td>
<td>173 ±5</td>
<td>1507 ±24</td>
<td>61 ±1</td>
<td>0.070 ±0.002</td>
<td>(4.0 ±0.2) x 10^{-4}</td>
<td>Yes</td>
</tr>
<tr>
<td>20</td>
<td>175 ±6</td>
<td>1558 ±26</td>
<td>59 ±1</td>
<td>0.067 ±0.001</td>
<td>(3.8 ±0.2) x 10^{-4}</td>
<td>Yes</td>
</tr>
</tbody>
</table>

The findings indicated higher loads for anodal discharges. They also emphasize that findings in experiments using saline do not accurately predict the physical phenomena seen in blood.

### TABLE 4
Gas formation as a function of polarity

<table>
<thead>
<tr>
<th>Medium</th>
<th>Anode (+)</th>
<th>Cathode (-)</th>
<th>Delivered energy (J)</th>
<th>Peak voltage (V)</th>
<th>Peak current (A)</th>
<th>Bubble volume/pulse (ml)</th>
<th>Bubble volume/J (ml)</th>
<th>Voltage waveform distortion</th>
</tr>
</thead>
<tbody>
<tr>
<td>Saline</td>
<td>Distal 6992A</td>
<td>8.5 cm disk</td>
<td>183 ±7</td>
<td>1833 ±40</td>
<td>56 ±1</td>
<td>0.111 ±0.003</td>
<td>(6.1 ±0.4) x 10^{-4}</td>
<td>Yes</td>
</tr>
<tr>
<td>Saline</td>
<td>8.5 cm disk</td>
<td>Distal 6992A</td>
<td>160 ±3</td>
<td>1374 ±25</td>
<td>65 ±1</td>
<td>0.067 ±0.006</td>
<td>(4.2 ±0.5) x 10^{-4}</td>
<td>Yes</td>
</tr>
<tr>
<td>Saline</td>
<td>Distal 6992A</td>
<td>Distal 6992A</td>
<td>215 ±4</td>
<td>2537 ±50</td>
<td>42 ±1</td>
<td>( + ) 0.082 ±0.002</td>
<td>(3.8 ±0.2) x 10^{-4}</td>
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</tr>
<tr>
<td>Blood</td>
<td>Distal 6992A</td>
<td>8.5 cm disk</td>
<td>187 ±6</td>
<td>1750 ±48</td>
<td>57 ±1</td>
<td>( - ) 0.034 ±0.002</td>
<td>(1.6 ±0.1) x 10^{-4}</td>
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</tr>
<tr>
<td>Blood</td>
<td>8.5 cm disk</td>
<td>Distal 6992A</td>
<td>158 ±6</td>
<td>1204 ±30</td>
<td>70 ±1</td>
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<td>(4.4 ±0.3) x 10^{-4}</td>
<td>Yes</td>
</tr>
<tr>
<td>Blood</td>
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<td>Distal 6992A</td>
<td>216 ±5</td>
<td>2430 ±93</td>
<td>44 ±2</td>
<td>( + ) 2.30 ±0.10</td>
<td>(106 ±7) x 10^{-4}</td>
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</tr>
<tr>
<td>Blood</td>
<td>6F, 2 mm Cu</td>
<td>8.5 cm disk</td>
<td>176 ±3</td>
<td>1733 ±28</td>
<td>61 ±1</td>
<td>( - ) 0.057 ±0.001</td>
<td>(2.6 ±0.1) x 10^{-4}</td>
<td>Yes</td>
</tr>
<tr>
<td>Blood</td>
<td>8.5 cm disk</td>
<td>6F, 2 mm Cu</td>
<td>140 ±5</td>
<td>1112 ±35</td>
<td>74 ±1</td>
<td>3.70 ±0.15</td>
<td>(210 ±12) x 10^{-4}</td>
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<tr>
<td>Blood</td>
<td>6F, 2 mm Cu</td>
<td>6F, 2 mm Cu</td>
<td>196 ±9</td>
<td>2320 ±42</td>
<td>50 ±1</td>
<td>( + ) 2.97 ±0.15</td>
<td>(152 ±15) x 10^{-4}</td>
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</tr>
<tr>
<td>Blood</td>
<td>Distal 6992A</td>
<td>8.5 cm disk</td>
<td>171 ±11</td>
<td>2098 ±103</td>
<td>55 ±1</td>
<td>( - ) 0.048 ±0.002</td>
<td>(2.4 ±0.2) x 10^{-4}</td>
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<tr>
<td>Blood</td>
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<td>Distal 6992A</td>
<td>174 ±7</td>
<td>1835 ±47</td>
<td>62 ±1</td>
<td>3.30 ±0.15</td>
<td>(193 ±21) x 10^{-4}</td>
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<td>(argon)</td>
<td>(argon)</td>
<td>(argon)</td>
<td>(argon)</td>
<td>(argon)</td>
<td>(argon)</td>
<td>(argon)</td>
<td>(argon)</td>
</tr>
</tbody>
</table>

Inter-electrode distance was 10 cm and energy setting 200 J in each case.
served in a room air atmosphere. Resistance to current flow, however, was slightly higher for either polarity. Delivered energy was similar for either polarity in an argon atmosphere when compared with a room air atmosphere (table 4).

Table 5 demonstrates the composition of the gas generated from discharges performed in a standard room air atmosphere and in an argon atmosphere for 30 min. The type of gas generated in a room air atmosphere in a blood medium (table 5) was a complex composition for either polarity discharge that could not have been caused by simple electrolysis. Diatomic nitrogen (N₂) was a large constituent of the gas bubble, regardless of whether the pulse was anodal or cathodal. This observation suggests that nitrogen dissolved from room air was expressed or squeezed from solution by the shock wave to form the primary component of the gas bubble. Diatomic oxygen (O₂), carbon dioxide (CO₂), argon, and diatomic hydrogen (H₂) are other gases that are dissolved in blood that also were evident in the gas generated from the procedure. Additionally, water vapor, monatomic nitrogen, and monatomic oxygen were present in small quantities and probably arose from thermal changes incurred in the arc and shock wave. Although all of these gases were present with either polarity pulse performed in a room air atmosphere, the percent composition differed slightly for the two polarity discharges.

In the presence of an argon atmosphere (table 5) applied for 30 min over blood, argon was a larger component of the gas composition of the bubbles, but in general, composition of the bubbles was similar to that of bubbles generated in a room air atmosphere. This increase in argon composition, even after a brief application of an argon atmosphere, provides further evidence that dissolved gases form a major part of the gas bubble.

Furthermore, defibrillator pulses applied to saline in a room air atmosphere (table 5) resulted in bubbles of a gas composition similar to that of bubbles observed in experiments performed in blood. The differences in percent composition of N₂, O₂, and CO₂ in particular probably reflect the effect of hemoglobin binding and intrinsic CO₂ production found in blood but not in saline. Regardless of these differences between saline and blood, the findings in saline are incompatible with simple electrolysis and add further support to the hypothesis that gas generated with this procedure is primarily from extrusion of dissolved gases from solution by a high-pressure shock wave. The mechanisms responsible for gas generation and the extent to which some of these gases derive from electrolysis will be discussed.

Discussion

From this study in vitro, we can conclude that electrode polarity predominantly, but also defibrillator energy and electrode surface area, significantly affect gas formation during delivery of damped sinusoidal discharges. Interelectrode distance, on the other hand, has a less pronounced effect on gas formation. In addition, this study has shown that the process of bubble formation is associated with a rapid alteration in the voltage waveform, which derives from impedance changes near the electrode and is dependent on the physics of spark formation in liquids. Furthermore, mass spectroscopy showed that there are multiple atomic and molecular components of the gas bubble, which excludes simple electrolysis as the only mechanism of bubble genesis. More likely, gas derives from high-pressure shock waves generated from the electric pulse. Together with previous work, these findings provide insight into the physics responsible for the problem and suggest ways to minimize gas formation in vivo during catheter-mediated electric pulses to cardiac tissue.

Each parameter evaluated in this study was chosen because of past or potential clinical application of the catheter technique for ablation of arrhythmogenic tissue with defibrillator pulses. Energy as high as 360 J, for example, was chosen because these high levels (or higher) have been used before on patients. The

<table>
<thead>
<tr>
<th>Anode (+)</th>
<th>Cathode (-)</th>
<th>Atmosphere</th>
<th>Medium</th>
<th>N₂</th>
<th>N</th>
<th>O₂</th>
<th>O</th>
<th>CO₂</th>
<th>H₂O</th>
<th>Ar</th>
<th>H₂</th>
</tr>
</thead>
<tbody>
<tr>
<td>Distal 6992A</td>
<td>8.5 cm disk</td>
<td>Room air</td>
<td>Blood</td>
<td>71.82</td>
<td>2.85</td>
<td>8.98</td>
<td>7.65</td>
<td>7.73</td>
<td>0.81</td>
<td>0.17</td>
<td>0.0000150</td>
</tr>
<tr>
<td>8.5 cm disk</td>
<td>Distal 6992A</td>
<td>Room air</td>
<td>Blood</td>
<td>65.21</td>
<td>5.02</td>
<td>17.63</td>
<td>5.66</td>
<td>5.13</td>
<td>0.54</td>
<td>0.81</td>
<td>0.0000999</td>
</tr>
<tr>
<td>Distal 6992A</td>
<td>8.5 cm disk</td>
<td>Ar (30 min)</td>
<td>Blood</td>
<td>71.60</td>
<td>4.48</td>
<td>14.35</td>
<td>3.81</td>
<td>3.57</td>
<td>0.37</td>
<td>1.81</td>
<td>0.000069</td>
</tr>
<tr>
<td>8.5 cm disk</td>
<td>Distal 6992A</td>
<td>Ar (30 min)</td>
<td>Blood</td>
<td>66.03</td>
<td>4.37</td>
<td>18.68</td>
<td>4.70</td>
<td>3.86</td>
<td>0.40</td>
<td>1.96</td>
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<tr>
<td>Distal 6992A</td>
<td>8.5 cm disk</td>
<td>Room air</td>
<td>Saline</td>
<td>42.94</td>
<td>1.87</td>
<td>46.96</td>
<td>5.26</td>
<td>0.13</td>
<td>1.61</td>
<td>1.24</td>
<td>0.000019</td>
</tr>
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<td>8.5 cm disk</td>
<td>Distal 6992A</td>
<td>Room air</td>
<td>Saline</td>
<td>33.42</td>
<td>2.68</td>
<td>55.88</td>
<td>5.62</td>
<td>0.11</td>
<td>1.56</td>
<td>0.72</td>
<td>0.000016</td>
</tr>
</tbody>
</table>
spectrum of energies studied from 360 J down to 5 J, except for the 5 and 10 J levels that are at the lower output limit of standard defibrillators, have also been used clinically. The Medtronic 6992A lead was chosen because of its structural integrity, which can tolerate high-energy pulses. Standard pacing and recording catheters were not tested because of their inherent dielectric weaknesses15,16 that can lead to leakage currents, which in turn would obfuscate the findings. Given the lack of a variety of high-dielectric strength clinical catheters suitable for study, copper electrodes were chosen as an approximation of standard No. 6F, 7F, and 8F size “catheters” to evaluate the effect of surface area on gas formation. An electrode length of 2 mm was chosen for study because it is a standard electrocatheter length for the tip electrode and allows for focused energy delivery. Electrode lengths of 5 and 10 mm, in conjunction with larger electrode diameter, were chosen as possibly practical means of increasing electrode surface area.

Interelectrode distances of 1 and 5 cm were studied because they were considered likely intracardiac anode-cathode separations. One centimeter spacing might apply, for example, if the anode and cathode were on the same catheter. A 5 cm spacing might be used when anode and cathode are on opposite sides of the ventricular septum. The 10 and 20 cm interelectrode spacings approximate the distance between an intracardiac and an anterior or posterior thoracic surface electrode. Polarity was studied because basic electrolytic and physical principles suggest that positive and negative charges can exert different influences on most solutions. Experiments were performed in anticoagulated fresh bovine blood as a readily available approximation to human blood. Saline baths were used in some experiments to help delineate mechanisms of bubble formation as well as to demonstrate the failure of saline to approximate events in blood. Argon atmospheres were used as an additional means of gaining insight into the mechanisms of gas formation.

The most important influence on gas production is polarity. A positive “ablating” electrode generates considerably more gas volume, nearly 50-fold, for a 200 J pulse setting than does a negative “ablating” electrode (3.30 vs 0.07 ml) in bovine blood (table 4). This difference is great and not easily explainable on the basis of electrolysis. In fact, even in absolute terms, the smaller volume of gas observed with the cathodal pulse cannot be explained by electrolysis.

For current delivered to water, electrolytic principles suggest that oxygen (O₂) should collect at the positive electrode and hydrogen (H₂) should collect at the negative electrode.20 However, our findings of multiple gases at both poles imply that the events taking place are more complicated than can be explained by electrolysis alone. The presence of N₂ in particular indicates that a considerable portion of the gas observed during the pulse arose from air that was dissolved in the blood. It is likely that N₂ as well as other gases were forced out of solution by a shock wave. This mechanism is at least theoretically possible, given that all the gases observed (except monatomic nitrogen and oxygen, which will be discussed later) are dissolved in sufficient quantities at atmospheric pressure to account for much of the bubble volume observed in this study.31 The concept that the gas generated by the pulse derives from dissolved air is further supported by the observation that under an argon atmosphere applied for 30 min, the percent of argon in the gas sample increased (table 5). This provides additional evidence that the bubble’s gaseous constituents arose, at least in part, from dissolved gases. Furthermore, the presence of the same gases, especially N₂, from samples collected in saline (table 5) also provides evidence that the bubbles arise from dissolved air and not from blood components per se. The difference in percent composition of the various gases generated in saline compared with those generated in blood is based on the considerable difference between the two liquid mediums. O₂ comprises a larger percentage of the gas generated in saline than of that generated in blood, most likely because hemoglobin binds O₂ in blood. CO₂ is more abundant in blood as a consequence of oxidative metabolism and therefore is more prominent in gas samples collected from experiments performed in blood. N₂ probably forms a larger percentage of the gas seen in the experiments with blood because it is the most common unbound gas dissolved in blood.32 In contrast, saline has approximately 1.6 times as much O₂ as N₂ dissolved per 100 ml at 25°C.31 Finally, it is important to note that within a particular medium differences in gas content can also occur due to polarity. These differences may arise from differences in effect on a particular gas by the arc and shock wave and from differences in electrolytic gas production.

To emphasize the role of shock waves on gas production, one must consider, in contrast, the role of electrolysis. Depending on electrode polarity, either H₂ or O₂ will collect at the “ablating” electrode in amounts predicted by Faraday’s law, which correlates the amount of charge delivered to the volume of gas generated (96,500 Coulombs will generate 1 mole or 22.4 liters of H₂ gas and 0.5 mole or 11.2 liters of O₂
We can approximate the amount of charge delivered for a 200 J anodal damped sine-wave pulse from the duration of the pulse and the measured peak current. Assuming a half-sine wave configuration for the defibrillator waveform, total charge delivered can be estimated from the following equation:

\[
\text{Charge} = 0.63 \times \text{peak current (Amps)} \times \text{pulse duration (sec)}
\]

For a 200 J stored anodal pulse (187 J delivered, 57 Amps peak current from table 4) using the distal 6992A Medtronic electrode, the charge delivered is 0.1436 coulombs. For a cathodal pulse, the same method of calculation yields a delivered charge of 0.1764 coulombs. By Faraday’s law, this calculation theoretically translates to 0.017 ml of O₂ for anodal pulses and 0.041 ml of H₂ for cathodal pulses. In neither case would electrolysis account for either the volume or the percent composition of the gas observed in our study.

On purely theoretical grounds, the amount of H₂ expected with cathodal pulses is actually less than observed. This finding may be due to several factors. One explanation may be that not all of the charge is transferred to water molecules. Part may be absorbed by proteins, red cell membranes, etc. Another explanation may be that with cathodal pulses part or all of the H₂ gas generated by electrolysis is combusted with dissolved oxygen during the arc. It is also conceivable that none of the H₂ gas collected in our samples was produced from electrolysis, but rather originated entirely from physiologic quantities of H₂ dissolved in the blood that were expressed from solution during the shock wave. Furthermore, shock waves may generate H₂ by dissociating H₂O. These latter two explanations would account for the presence of H₂ in gas generated with anodal pulses as well as with cathodal pulses.

Similar problems arise when trying to understand the volume of O₂ and the percent O₂ concentration observed on the basis of electrolysis. With the use of data from tables 4 and 5, simple arithmetic calculations demonstrate that actual O₂ volume with the anodal pulse (0.296 ml) is greater than would be expected from electrolysis alone (0.017 ml). This finding also implies that a substantial portion of the O₂ arises as a consequence of a shock wave that expresses dissolved O₂ from solution. The presence of O₂ in bubbles generated with cathodal pulses (which should generate H₂ according to electrolysis theory) confirms this contention. (The volume of O₂ generated from cathodal pulses is 17.63% of 0.070 ml or 0.0123 ml.)

These findings do not indicate that electrolysis is an insignificant factor in the development of the phenomena observed. In fact, shock-wave generation is precipitated in part by electrolysis. A clue to the manner in which shock waves arise can be gleaned first from changes observed in the voltage waveform. We have noted distortions or rises in what should otherwise be a smooth Edmark voltage waveform during those pulses resulting in arcs (table 1, figure 3). These irregularities in the waveform contour indicate changes in impedance after the initial gas formation that results from electrolysis. This process probably begins with the electrolytic decomposition of H₂O to H₂ and O₂ and with heating of water. The quantities of gas generated by electrolysis are quite small and serve only as a catalyst to subsequent events. Apparently, when enough gas is generated by electrolysis to create a bubble large enough to envelop and insulate the electrode from the blood, current flow to the surrounding blood is transiently interrupted and impedance rises. Despite the rise in impedance, current continues uninterrupted to the electrode because of the inertia inherent in the Physio-Control defibrillator inductor, leading to a rise in voltage between the bubble-insulated electrode and the surrounding blood. Because of this bubble insulator, an overvoltage is generated between electrode and blood as the defibrillator pulse continues to be delivered to the electrode. As the electric field strength increases, more and more electrons enter the bubble that surrounds the electrode until an electron “avalanche” occurs. Eventually, electron density is strong enough to sustain an arc. With sufficient electron density and current arcing between the electrode and the surrounding blood, a flash occurs that can result in an enormous temperature rise in the bubble (as high as 6000°C). Such high temperatures result in a very rapidly expanding bubble volume and an extremely rapid change in the thermodynamic state of the gas. By definition, this is a shock wave. The shock wave thus generated can be as high as 50,000 atmospheres, although it is usually less (10 to 20 atmospheres).

Although shock waves can arise from rapid bubble expansion, they may also occur as a result of “bubble collapse.” Bubble collapse occurs after gas bubble overexpansion, which leads to a fall in bubble pressure below ambient pressure. As the momentum of the expanding bubble fades and the ambient pressure exceeds intrabubble pressure, the volume of the bubble begins to contract. This contraction in bubble volume may be extreme and precipitate complete collapse of the bubble. On bubble collapse, a shock wave may again be generated. In some instances, one may actual-
ly observe successive shock waves as a consequence of bubble ringing.36 Regardless of the mechanism of shock-wave generation, it is this phenomenon that we believe leads to extrusion of gas from solution. Additionally, it should be noted that the high temperatures and high pressures also account for the presence of monatomic nitrogen and monatomic oxygen.44, 45 Both events can dissociate N\textsubscript{2} and O\textsubscript{2} into their monatomic states.

Knowledge of the physical processes leading to arcing and shock-wave formation during defibrillator discharges to blood allows us to understand why anodal pulses generate nearly 50 times the gas volume of cathodal pulses. The reason probably relates to the manner in which arc formation occurs with each polarity pulse. For cathodal pulses, electrons are emitted from the catheter electrode. For anodal pulses, electrons are emitted from the surrounding blood. Because electrons can more easily be extracted from a relatively sharp point (i.e., an electrode) rather than from a larger, more diffuse area (i.e., the blood at the margins of the gas bubble — a more or less spherical structure) for any given electric field strength, cathodal pulses are more likely to arc with smaller electric field strengths than are anodal pulses. In addition to spatial considerations, there is empirical evidence that electric gaseous breakdown occurs at lower field strengths for cathodal pulses in all but very low gas pressure environments (i.e., vacuums).39 Furthermore, the relative spark-breakdown strength of H\textsubscript{2} is less than that of O\textsubscript{2},46 also indicating that a smaller electric field strength will lead to arcing in cathodally generated gas (i.e., H\textsubscript{2}) more easily than in anodally generated gas (i.e., O\textsubscript{2}). Consequently, because anodal pulses require higher electric field strength (i.e., voltages) to arc for a given energy (tables 1 and 4), a larger shock wave is generated. Früngel\textsuperscript{35} has confirmed this concept and suggested that as much as 50% of the electrical energy supplied by the spark (note: not total delivered energy) is converted into mechanical energy. The relationship between electric field strength and energy is more apparent when one considers the equation governing the electrical energy of the arc, as follows:

\[ E = \frac{1}{2} CV^2 \]

where E is the energy in joules, C is the capacitance (farads), and V is the electric field strength (volts) across the gas bubble (not the entire system).35 Because energy varies as the square of the field strength, small increases in the voltage needed to precipitate electrical breakdown of the bubble result in consider-

able increases in the energy of the arc. Therefore, anodal pulses result in higher voltages and the amount of energy in the arc associated with an anodal pulse is greater than that associated with a cathodal pulse. Consequently, more electrical energy is available in an anodal pulse for translation into mechanical energy (i.e., high-pressure fronts), which should, in turn, lead to more gas being expressed from solution.

Another finding in this study is that saline does not approximate blood well enough to predict the physical phenomena associated with catheter-mediated defibrillator pulses (compare saline and blood data in table 4), especially when considering the effect of anodal pulses. An anodal pulse in blood generated approximately 30 times the bubble volume generated by the same anodal pulse in saline. Cathodal pulses were essentially equivalent in their ability to generate gas in either medium. Why the difference? Although there are no theories to explain this phenomenon, it is possible to venture a hypothesis based on related phenomenon. We know, for example, that laser-induced liquid breakdown that results in rapidly expanding and collapsing bubbles leading to shock waves are dependent on “impurities” in the liquid that is subjected to the laser pulses.42, 47 These impurities tend to absorb energy and form the center about which a plasma develops that eventually leads to arcing, shock-wave generation,42 and gas formation.48 By a parallel reasoning process, one might state that blood contains more “impurities” (cells, macroproteins) than saline and would therefore be more likely to concentrate energy at these niduses.

A more immediate explanation for the differences between anodal gas formation in blood and saline may involve the electric field strength needed to breakdown the gas initially generated by electrolysis. Cobine\textsuperscript{49} has noted that breakdown voltage may rise considerably with impurities on the cathode. Impurities in the case of blood could be anything that limits electron flux from the cathode. In the case of an anodal catheter discharge, the cathode becomes the surface of the gas bubble created by electrolysis. As explained earlier, the bubble’s structural configuration alone (i.e., a large sphere rather than the relative point configuration of the electrode), makes electron emission more difficult. Add to this the “impurities” of lipids, red cells, macroproteins, etc. that coat the surface of the bubble, and electron extraction becomes more difficult, requiring higher electric field strength across the bubble (again, not across the whole system). As a result, more electric energy can be translated to mechanical energy with anodal pulses delivered in blood than with anodal pulses.
pulses delivered in saline. (It is important to note that this discussion of energies and field strengths refers to localized phenomenon within and across the gas bubbles generated by electrolysis; it does not refer to the energies and voltages measured in this study that reflect more global events throughout the entire circuit.)

Given the findings of this study and the present technique for catheter-mediated electrical ablation of arrhythmogenic tissue, what can be done clinically to limit gas formation yet still produce the desired elimination of the arrhythmogenic focus? Clearly, in addition to use of cathodal pulses, one method is to use lower energy. We are limited in this approach, however, by our understanding of the relationship of energy used (or voltage, current, or additional as yet undefined factors) to the amount of tissue damaged. If energy is decreased to minimize bubble formation, adequate tissue injury may not occur. Unfortunately, there are few data regarding the energy necessary to ablate a discrete focus of myocardial tachycardia. In one clinical report, as many as six consecutive 300 J pulses through a No. 6F catheter were used for attempted ablation of a focus of ventricular tachycardia. This amount of energy could generate 0.6 ml of gas in our preparation in vitro, even if cathodal pulses were used (and could generate over 21 ml if anodal pulses were used). If catheter-mediated pulses are to be considered free from potential gas emboli, it appears that smaller energies should be employed.

Lerman et al. have demonstrated that cathodal pulses of 50 to 100 J delivered to healthy canine myocardium result in 1.7 to 2.5 g of necrotic cardiac muscle. This volume of damaged left ventricular myocardium may be near the amount needed to prevent arrhythmogenesis if one is treating a very discrete tachycardia focus and if one makes the substantial assumption that the damage will be equivalent in diseased or scarred myocardium. Given these limitations, it may be possible to compare these findings with catheters with better defined cryosurgical techniques to determine whether 50 to 100 J defibrillator pulses may be clinically useful. As few as three cryolesions have been used during surgery for destruction of discrete arrhythmogenic foci in patients with recurrent sustained monomorphic ventricular tachycardia, the sites of origin of which are well localized during endocardial mapping. Work in dogs has shown that a typical cryolesion will result in a volume of injured tissue of approximately 2.1 ml. Comparing the volume of myocardium injured with a cryolesion to that injured with an electrical pulse on the order of 50 to 100 J, we might find that three adjacent 50 J pulses are adequate to alter a very discrete arrhythmogenic focus. Consequently, use of energies in this range would generate smaller gas volumes and minimize concerns of bubble emboli. With the use of smaller energy pulses, one may also generate smaller shock waves and therefore decrease concerns regarding barotrauma.

An additional means to decrease gas formation would be the use of electrodes of larger surface areas (table 2). This could be accomplished by increasing electrode diameter and/or length to a limited degree. One must be aware, however, that such an alteration in size of the electrode to minimize gas formation could conceivably result in a less favorable, or at least a more unpredictable, outcome with respect to tissue ablation. For example, there may be a significant alteration in the amount of injured tissue resulting from any particular delivered pulse current when an electrode of a larger surface area is used. Theoretically, tissue injury may be either more or less than that occurring with a standard electrode, depending on the trade-off between lower current density and larger area of electrode-tissue contact as well as on the relative effects of electrical and barotraumatic injury. No catheter data, however, are available regarding the effect of electrode surface area on tissue injury. Nevertheless, we do know from previous studies of defibrillation that larger paddles are less likely to injure the myocardium for any particular energy used. This finding suggests that tissue injury may be limited for any particular energy if electrodes of larger surface areas are used to decrease gas formation.

An increase in the diameter or length of the electrode may present other difficulties as well. In particular, catheter maneuverability would be limited compared with that of the standard No. 6F catheter that is presently used. Furthermore, the “discreteness” of any attempt might be lost with longer or wider electrodes. One can state, therefore, that the effect of changes in surface area on cardiac injury as required for catheter-mediated ablation of arrhythmogenic tissue is not yet certain. Nonetheless, it is clear that increases in surface area will produce decreases in bubble volume.

The clinical consequences of bubble formation during catheter-mediated pulses, particularly in the left heart, require close scrutiny, given the potential risks of embolization. Theoretically, gas bubbles from catheter-mediated pulses might embolize to the coronary arteries or to the cranial circulation. Their clinical significance would depend on the quantity generated and on the disease status of the vessel involved. The volume of gas formed, for example, with a 200 J anodal pulse delivered to the Medtronic 6992A distal elec-
trode (table 1) is on the order of 3.5 ml, a substantial gas embolus by any measure and clearly undesirable. What about the smaller volumes observed with cathodal pulses — are they dangerous? The answer may be placed in perspective by reviewing the results of canine studies in which air was injected directly into the coronary artery and by comparing these findings with the calculated volume of air commonly seen with accidental air injection during coronary arteriography in man — normally an innocuous occurrence. In dogs, Goldfarb and Bahnson9 showed that the apparently trivial amount of 0.1 to 0.2 ml of air injected directly into the left anterior descending coronary artery was sufficient to result in myocardial injury. As a point of reference, one might consider the typical air bubble injected accidentally during coronary arteriography. Practical experience tells us that this accident seldom has clinical significance. The bubble appears to traverse the arterial system quickly. However, the volume of a gas bubble injected accidentally during coronary arteriography and the volume of gas injected during Goldfarb’s study are considerably different. A normal epicardial coronary artery, midway along its course, has a diameter of 2 mm.56 With the use of the formula for the volume of a sphere, a bubble of 2 mm diameter has a volume of 0.0042 ml. A volume of gas of 0.1 ml associated with a 300 J cathodal pulse, for example (table 1), would contain the equivalent of 28 bubbles, each with a 2 mm diameter. The question of whether or not 0.1 ml of gas causes clinical injury obviously cannot be easily dismissed, especially with delivery of multiple 300 J pulses. Of the studies in which left ventricular defibrillator pulses were used, each has reported delivery of multiple pulses and a significant rise in creatine kinase-MB levels.4,5-8 Although this rise in enzyme level is more likely a consequence of electrical or barotraumatic55, 56, 53, 54 injury rather than anything else, a component of myocardial injury related to bubble embolization cannot be excluded. The concern over systemic and particularly cerebral embolization is similar. From the present literature, however, there is no evidence to suggest that cerebral injury has occurred. In any event, these considerations emphasize the potential embolic risks of delivery defibrillator pulses to the left ventricle and suggest that clinical applications proceed in a measured fashion.

In addition to the concern over gas emboli, there is at least equal concern that the high-pressure shock wave associated with this technique may result in barotraumatic cardiac rupture. As stated earlier, underwater capacitor discharges of similar energies have resulted in near-field barometric changes on the order of thousands of millimeters of mercury for short durations.55, 56 The fact that these pressure changes can occur with defibrillator discharges as well is confirmed in figure 2. The consequences of such pressure rises have been demonstrated in previous work in vivo.55, 54 Histologic findings in canine studies substantiate the presence of cardiac barotrauma during delivery of catheter-mediated defibrillator pulses to the coronary sinus. These studies showed rupture of the coronary sinus elastica in each dog, and in some animals, cardiac rupture, indicating that caution should be exercised with the procedure if applied to man.

In summary, these studies have shown that electrode polarity, and to a lesser extent, pulse energy, electrode surface area, and interelectrode spacing, affect bubble formation during catheter-mediated delivery of defibrillator pulses to a preparation of bovine blood in vitro. The volume of gas generated may be substantial, which could theoretically result in a clinically significant embolus. The occurrence of gas formation is heralded by alterations in the defibrillator voltage waveform, which is a reflection of the factors leading to shock-wave generation and to the subsequent extrusion of dissolved gases from solution.

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