The use of an exponential biphasic waveform has been shown to be superior to an exponential monophasic waveform in external1 as well as internal2,3 defibrillation. Use of a single capacitor for generating exponential biphasic waveforms poses several limitations, however. Increasing the phase-1 duration may not generate the optimal phase-2 leading-edge voltage, capacitance, and impedance. Therefore, the optimal single-capacitor biphasic waveform may not generate the best combination of phase 1 and phase 2 compared with the optimal combination of 2 separate capacitors in both phases.

In a previous report,4 we found that extending the phase-1 tilt of a biphasic waveform from 30% to 70% lowered the defibrillation threshold (DFT) energy when a 60-μF phase-1 capacitor was used. This finding suggested that the optical tilt may be >70%. Second, that previous report4 compared the use of a phase-2 capacitor (20 μF) that was smaller than the phase-1 capacitor with the use of same-size capacitors in both phases. The results indicated that the use of a smaller capacitor in phase 2, when charged to the same leading-edge voltage as phase 1, yielded the optimal DFT energies.

External defibrillation efficacy is influenced by multiple factors, such as electrode pad size,5,6 shock waveform,1,7–10 and sequential shocks.11 Because capacitor size may also play a role in determining external defibrillation efficacy,12 optimizing capacitor sizes may contribute to maximizing defibrillation efficacy. Theoretical models, assuming a 50-Ω impedance, have suggested that capacitors in the 30- to 60-μF range may provide the optimal capacitor for phase 1 of a biphasic waveform.13,14 Use of a smaller capacitor in a fully discharging capacitor waveform may be particularly important, because larger capacitors could result in exceedingly long pulse widths when the defibrillation shock is applied through a higher-impedance pathway. For example, for trans-thoracic impedance of 80 to 100 Ω,8–10,15 discharging a 120-μF capacitor to 95% tilt would take >30 ms.

The purpose of this study was to assess the optimal capacitor sizes (phase 1 and phase 2) for a fully discharging (95% tilt) biphasic waveform in the above-discussed range of 30, 60, and 120 μF that could be implemented in a standard external defibrillation device.

Background—Phase-2 voltage and maximum pulse width are dependent on phase-1 pulse characteristics in a single-capacitor biphasic waveform. The use of 2 separate output capacitors avoids these limitations and may allow waveforms with lower defibrillation thresholds. A previous report also suggested that the optimal tilt may be >70%. This study was designed to determine an optimal biphasic waveform by use of a combination of 2 separate and fully (95% tilt) discharging capacitors.

Methods and Results—We performed 2 external defibrillation studies in a pig ventricular fibrillation model. In group 1, 9 waveforms from a combination of 3 phase-1 capacitor values (30, 60, and 120 μF) and 3 phase-2 capacitor values (0=monophasic, ¼, and 1.0 times the phase-1 capacitor) were tested. Biphasic waveforms with phase-2 capacitors of ¼ times that of phase 1 provided the highest defibrillation efficacy (stored energy and voltage) compared with corresponding monophasic and biphasic waveforms with the same capacitors in both phases except for waveforms with a 30-μF phase-1 capacitor. In group 2, 10 biphasic waveforms from a combination of 2 phase-1 capacitor values (30 and 60 μF) and 5 phase-2 capacitor values (10, 20, 30, 40, and 50 μF) were tested. In this range, phase-2 capacitor size was more critical for the 30-μF phase-1 than for the 60-μF phase-1 capacitor. The optimal combinations of fully discharging capacitors for defibrillation were 60/20 and 60/30 μF.

Conclusions—Phase-2 capacitor size plays an important role in reducing defibrillation energy in biphasic waveforms when 2 separate and fully discharging capacitors are used. (Circulation. 1999;100:826-831.)

Key Words: defibrillation • death, sudden • ventricles

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Methods

The use of experimental animals in this study was approved by the Animal Research Committee of the Cleveland Clinic Foundation and conformed to the recommendations of the American Heart Association on Research Animal Use.

Study Preparation and Surgical Procedure

Anesthetized pigs were used in this study. Animal preparation and surgical procedures have been described in detail in previous publications. Briefly, the swine were intubated with auffed endotracheal tube and ventilated with room air supplemented with oxygen through a Drager SAV respirator (North American Drager), which was adjusted as needed to maintain normal arterial blood gases. A transvenous unipolar defibrillation lead (model 497, Intermedics Inc) was inserted into the right ventricular (RV) apex through the right external jugular vein under fluoroscopy. Three adhesive pad electrodes, each with a surface area of 75 cm², were applied to the right upper pectoral, the left upper pectoral, and the cardiac apical area on shaved skin.

Defibrillation Protocol

The 3 adhesive pad electrodes and the RV apex defibrillation lead were connected to an external defibrillator custom-made by Survialink Corp. The function of this external defibrillator was described in detail in our previous publication. Ventricular fibrillation (VF) was induced by delivery of 60-Hz AC (15 V) for 3 seconds through the RV apex defibrillation lead and the left pectoral pad electrode. After VF sustained for 10 seconds, 1 of the test waveforms was delivered between the right upper pectoral pad electrode and the apical pad electrode at the end-respiration phase. The apical pad electrode was the anode for phase 1 of the biphasic waveforms. If defibrillation failed for the test shock waveform, a rescue shock (450 to 900 V) was delivered between the RV defibrillation lead and the left upper pectoral pad electrode. A recovery period of ≥3 minutes was allowed between each episode of VF. VF was not reinitiated until heart rate and blood pressure returned to the preshock values.

Defibrillation Waveforms

Two groups of experiments were performed for this report. In group 1, 3 phase-1 capacitor sizes (30, 60, and 120 μF) and 3 phase-2 capacitor sizes (0=monophasic, ⅓ of, and equal to phase-1 capacitor) were tested for a total of 9 different test shock waveforms. These 9 waveforms consisted of 3 exponential monophasic and 6 exponential biphasic waveforms, as illustrated in Figure 1A.

In group 2, a more detailed assessment of optimal phase-2 capacitor was performed using phase-1 capacitor sizes of 30 and 60 μF. Because of concerns regarding the potential length of phase-1 distal when 120-μF capacitors were used and the experimental limitation of the number of waveforms that can be tested in a single experiment, the 120-μF phase-1 capacitor was not evaluated. Five phase-2 capacitor sizes (10, 20, 30, 40, and 50 μF) were used in combination with the 30- and 60-μF phase-1 capacitor. Ten different exponential biphasic waveforms were tested in this group to evaluate defibrillation efficacy, as shown in Figure 1B.

Evaluation of Defibrillation Efficacy

Defibrillation efficacy of each waveform was estimated by \( V_{50} \), defined as the leading-edge voltage of the waveform associated with a 50% likelihood of successful defibrillation. \( V_{50} \) was measured by a previously described Bayesian estimation technique. Ten defibrillation tests were performed for each waveform to evaluate \( V_{50} \). The first shock phase-1 leading-edge voltage was 1650 V in all waveforms. Sequential step changes in voltage were 350, 200, 150, 150, 100, 100, 50, 50, and 0 V. These steps in voltage change were either positive or negative, depending on failure or success in defibrillation of the preceding shock, respectively. \( E_{50} \) was the energy stored in the capacitors calculated from these \( V_{50} \) values.

In each animal, the effect of the first shock (1650 V) was tested first in a randomized order for all waveforms. Then, second shock voltages for the waveforms were determined, based on the result of the first, and were applied in random order as well. Thereafter, by the same procedure, the third through 10th shocks were delivered.

Statistical Analysis

Data in all DFT parameters were expressed as mean values ± SD. Repeated-measures 1-way ANOVA was used to compare DFT parameters in group 1 and group 2. Pairwise comparisons of the waveforms were made for each parameter, with the least significant difference test used in group 1 and group 2. The null hypothesis was rejected for \( P < 0.05 \).

Results

Group 1

A complete data set was obtained for 10 pigs (36 ± 2 kg). All parameters at DFT are shown in Table 1.

Figure 2 shows the \( E_{50} \) of stored energy and \( V_{50} \) of phase-1 leading-edge voltage for all waveforms. For waveforms using a 30-μF phase-1 capacitor, there was a trend toward lower \( E_{50} \) in the biphasic waveforms, but this difference did not reach statistical significance. However, there was a significant drop in leading-edge voltage for the biphasic waveforms, with the lowest voltage seen in the 30/30-μF waveform. In the waveforms using a 60-μF phase-1 capacitor, the optimal \( E_{50} \) and \( V_{50} \) were seen with use of the 20-μF phase-2 capacitor (\( P = 0.008 \) and \( P = 0.013 \), respectively). In the waveforms using a 120-μF phase-1 capacitor, the optimal \( E_{50} \) and \( V_{50} \) were observed with a phase-2 capacitor of 40 μF. Interestingly, using a phase-2 capacitor equal to phase 1 (120/120 μF) resulted in \( E_{50} \) and \( V_{50} \) that were significantly higher even compared with the corresponding monophasic (120-μF) waveform. Thus, the biphasic waveform with a smaller phase-2 capacitor than phase-1 capacitor reduced the \( E_{50} \) of stored energy when 60- and 120-μF capacitors were used in phase 1. For a phase-1 30-μF capacitor, the 30/30-μF
through the E 50 estimates did not reach statistical differences. These results suggest that a phase-2 capacitor of 20 to 40 μF may be optimal for a wide range of phase-1 capacitors.

**Group 2**

A complete data set was obtained for 10 pigs (33±4 kg). All parameters at DFT are shown in Table 2.

Figure 3 shows the results of this study show that DFT voltage and/or energy can be significantly lowered with the use of biphasic discharging capacitor waveforms compared with their corresponding monophasic waveforms. It is interesting to compare the effects of the phase-2 capacitor on the DFT. When a 30-μF phase-1 capacitor was used, the addition of a phase-2 capacitor (10 or 30 μF) lowered both DFT energy and voltage by ~40%. The lowest DFT was associated with the 30-μF phase-2 capacitor. With the 60- and 120-μF phase-1 capacitors, adding the corresponding 20- and 40-μF phase-2 capacitors had similar effects in lowering the DFT. However, when the corresponding phase-2 capacitors were made equal to the phase-1 capacitors, DFTs increased again, and for the 120-μF capacitors, biphasic thresholds were even higher than monophasic thresholds. These results would suggest that the optimal phase-2 capacitor for 30- to 120-μF phase-1 capacitors is in the range of 20 to 40 μF regardless of the capacitor used in phase 1. The results of these experiments showed that the capacitor combinations of 60/20 and 120/40 μF were clearly superior to the corresponding ones using the same capacitors for both phase 1 and phase 2 (60/60 and 120/120 μF). Perhaps because 30 μF was in the range of the optimal phase-2 capacitor, the difference between the 30/10- and the 30/30-μF waveforms was small.

The results shown in Table 1 reveal that the mean impedances in our experimental model were in the 40- to 50-Ω range. However, typical transthoracic impedances for human defibrillation are higher, 60 to 80 Ω, with values in the 80- to 100-Ω range seen occasionally. The use of a 120-μF capacitor may be problematic in this impedance range. Based on the pulse widths shown in Table 1, one can extrapolate that the use of the same 120-μF waveforms in humans can extend the pulse widths to the 20- to 30-ms range. Such long pulse widths are likely to be not as effective for defibrillation, because their duration will be well beyond the chronaxie of the defibrillation strength-duration curves. Thus, group 2 data shown in Table 2 were limited to a more detailed analysis of the 30- and 60-μF waveforms.

As illustrated in Figure 3, the pattern of changes in the DFT voltage and energy associated with changing phase-2 capac-
itors was different when the 30-μF and the 60-μF phase-1 capacitors were compared. For the 30-μF phase-1 capacitor, there was a clear minimum $V_{50}$ associated with the use of a 30-μF capacitor for phase 2. This minimum was reflected in the $E_{50}$ values as well. When the size of the phase-2 capacitor exceeded that of phase 1, the DFT values rose markedly. However, in the 60-μF phase-1 capacitor, the variation in the phase-2 capacitor did not have as clear a minimum as that shown for the 30-μF phase-1 data, perhaps partly because a phase-2 capacitor size that exceeded that of phase 1 was lacking. A minimum for both $E_{50}$ and $V_{50}$, however, was seen at a phase-2 capacitor value of $\approx 20$ to 30 μF. Like the observations in internal defibrillation,18,19 the results of this study showed that exponential biphasic waveforms with a very small capacitor (20 to 40 μF) in phase 2 were effective for external defibrillation. Furthermore, the 60-μF waveform had lower $V_{50}$ values than the 30-μF waveforms. We would postulate that the 60-μF biphasic waveform with a phase-2 capacitor of $\approx 20$ to 30 μF may be the optimal combination among the ones reported here for human external defibrillation. The 120/40-μF waveform had a lower-voltage DFT at similar-energy DFT but would probably generate excessively long pulses for the higher impedances.

**Fully Discharging Capacitor Waveform**

Because the tail of the defibrillation pulse may act to refibrillate the myocardium, an appropriate truncation of waveforms has been postulated to reduce DFT energy.13 However, previous publications20,21 have reported somewhat conflicting results of truncation in monophasic capacitor discharge waveforms. The effect of truncating phase 1 in biphasic waveforms has not been studied in detail for defibrillation with a dual-capacitor system. Our recent external defibrillation study1 showed that DFT energy decreased with phase-1 tilt, increasing from 30% to 70% when a separate phase-2 capacitor was used. This finding suggested that the optimal phase-1 tilt for dual-capacitor biphasic waveforms may be >70%. Despite previous experimental observations that truncation is important in monophasic waveforms, the data reported here raise the interesting possibility that such truncation may not play as significant a role in biphasic waveforms. The results with the 120-μF phase-1 waveform are an example of this possibility. Despite use of a full-tilt phase 1 that has relatively high $E_{50}$ and $V_{50}$ when tested as a monophasic waveform, the addition of a small phase-2 capacitor markedly decreased the DFT to values that were even better than those for the 30- and 60-μF capacitors.

**Changing Capacitance at Phase Reversal**

A recent external defibrillation study1 showed the superiority of single-capacitor biphasic waveforms over exponential monophasic waveforms. However, there are a few limitations in such a single-capacitor biphasic waveform. Maximizing

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### TABLE 2. DFT Parameters and Waveform Characteristics in Group 2

<table>
<thead>
<tr>
<th>Waveform</th>
<th>Capacitor Size in μF</th>
<th>Stored Energy, J</th>
<th>Phase-1 Leading-Edge Voltage, V</th>
<th>Phase-1 Impedance, Ω</th>
<th>Phase-1 Pulse Width, ms</th>
<th>Phase-2 Pulse Width, ms</th>
</tr>
</thead>
<tbody>
<tr>
<td>30/10</td>
<td>65±24†</td>
<td>1779±315</td>
<td>44±3</td>
<td>3.7±0.2</td>
<td>1.2±0.1</td>
<td></td>
</tr>
<tr>
<td>30/20</td>
<td>60±14†</td>
<td>1540±197</td>
<td>45±4</td>
<td>3.7±0.2</td>
<td>2.4±0.1</td>
<td></td>
</tr>
<tr>
<td>30/30</td>
<td>55±14‡</td>
<td>1346±167</td>
<td>45±3</td>
<td>3.8±0.2</td>
<td>3.7±0.2</td>
<td></td>
</tr>
<tr>
<td>30/40</td>
<td>93±42§</td>
<td>1597±342</td>
<td>45±3</td>
<td>3.7±0.2</td>
<td>4.9±0.3</td>
<td></td>
</tr>
<tr>
<td>30/50</td>
<td>143±51</td>
<td>1862±332</td>
<td>44±3</td>
<td>3.7±0.2</td>
<td>6.0±0.3</td>
<td></td>
</tr>
<tr>
<td>60/10</td>
<td>62±23†</td>
<td>1301±275</td>
<td>45±3</td>
<td>7.5±0.4</td>
<td>1.2±0.1</td>
<td></td>
</tr>
<tr>
<td>60/20</td>
<td>42±18*</td>
<td>996±224</td>
<td>46±3</td>
<td>7.6±0.4</td>
<td>2.5±0.1</td>
<td></td>
</tr>
<tr>
<td>60/30</td>
<td>43±16†</td>
<td>959±176</td>
<td>48±4</td>
<td>7.7±0.5</td>
<td>3.8±0.2</td>
<td></td>
</tr>
<tr>
<td>60/40</td>
<td>54±20‡</td>
<td>1026±192</td>
<td>48±5</td>
<td>7.7±0.5</td>
<td>5.0±0.3</td>
<td></td>
</tr>
<tr>
<td>60/50</td>
<td>59±19†</td>
<td>1027±165</td>
<td>47±5</td>
<td>7.7±0.5</td>
<td>6.2±0.4</td>
<td></td>
</tr>
</tbody>
</table>

*†P<0.05 vs 30/10, 30/20, 30/40, 30/50, 60/10 and 60/50.‡P<0.05 vs 30/10, 30/20, 30/40, 30/50 and 30/50.§P<0.05 vs 30/50.

All pairwise comparisons of phase-1 leading-edge voltage are significant ($P<0.05$) except for 30/10 vs 30/50, 30/20 vs 30/40, 30/30 vs 30/50, 60/10 vs 60/30, 60/20 vs 60/40, 60/20 vs 60/30, 60/30 vs 60/40, 60/20 vs 60/50, 60/30 vs 60/50, and 60/40 vs 60/50. All pairwise comparisons of phase-2 pulse width are significant ($P<0.05$) except among comparisons of same-size phase-1 capacitor waveforms. All pairwise comparisons of phase-2 pulse width are significant ($P<0.05$) except among comparisons of same-size phase-2 capacitor waveforms.

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![Figure 3. Mean and SD of $E_{50}$ of stored energy and $V_{50}$ of phase-1 leading-edge voltage for 10 biphasic waveforms in group 2. Left and right vertical axes show $E_{50}$ of stored energy (J) and $V_{50}$ of phase-1 leading-edge voltage (V), respectively; horizontal axis indicates each waveform.](image-url)
the phase-1 pulse width may not generate the optimal phase-2 leading-edge voltage or charge transfer, because phase-2 leading-edge voltage is dependent on phase-1 pulse width. In addition, several recent defibrillation studies have shown that a biphasic waveform with a smaller phase-2 capacitor can achieve lower defibrillation threshold (DFT) energy when a phase-1 capacitor of 60 \( \mu F \) is used. Therefore, optimal biphasic waveforms may be best generated with 2 separate capacitors, 1 for phase 1 and 1 for phase 2.

As shown in Figure 2, the addition of an appropriate phase-2 capacitor can improve the E\(_{50}\) of the biphasic waveform over that obtained with the corresponding monophasic waveform. However, the relative improvement of the E\(_{50}\) appears to be greater when larger capacitors are used in phase 1. Specifically, the biphasic waveforms of 30/10, 60/20, and 120/40 \( \mu F \) reduced E\(_{50}\) by 38%, 67%, and 81%, respectively, compared with their respective monophasic waveforms. Thus, the beneficial effects of adding a phase-2 capacitor to the monophasic waveform appear to depend on the capacitor size in phase 1. Although the monophasic 120-\( \mu F \) waveform had the highest E\(_{50}\) of the 3 capacitors, the greater improvement in E\(_{50}\) seen with the addition of phase 2 made the 120/40-\( \mu F \) waveform perform as well as the 60/20-, 30/10-, and 30/30-\( \mu F \) waveforms. Another interesting observation is the effect of having the same capacitor/pulse width in both phases of the biphasic waveform. Although the 120/120-\( \mu F \) waveform had a poor E\(_{50}\) compared with the 120/40-\( \mu F \) waveform, the difference was much less prominent when the 60/60-\( \mu F \) and the 60/20-\( \mu F \) waveforms were compared, and the difference was nonexistent when the 30/30-\( \mu F \) and the 30/10-\( \mu F \) waveforms were compared. This observation would suggest that the optimal capacitor for a phase-2 fully discharging waveform may be fairly constant, \( \approx 20 \) to 40 \( \mu F \), regardless of the phase-1 capacitor used. Because phase-2 capacitors were charged to the same voltage as phase-1 capacitors in these experiments, this observation would imply that the actual need for charge transfer in phase 2 may be less for higher-capacitor phase-1 waveforms where the leading-edge voltage is lower, even though the pulse width is longer, at least within the impedance range seen in our experimental preparation. Thus, the need for phase-2 charge transfer may be related to the leading-edge voltage of phase 1 regardless of the phase-1 capacitor.

**Applicability of the 120-\( \mu F \) Capacitor to Human External Defibrillation**

The impedance for external shock in the human chest is 60 to 80 \( \Omega \) and may exceed 100 \( \Omega \) in some cases. Conversely, the shock impedance of our pig model was \( \approx 40 \) to 50 \( \Omega \) in this study. Although the 120/40-\( \mu F \) biphasic waveform performed quite well, as evidenced by the data presented in Table 1 and Figure 2, its applicability to higher-impedance pathways may be problematic. The phase-1 pulse width of the 120-\( \mu F \) waveforms tested here was \( \approx 14 \) ms with 40-\( \Omega \) impedance. For clinical circumstances, this phase-1 pulse width may become 20 to 30 ms because of the higher impedance and may be even higher in high-impedance cases. Such a long pulse width would most likely be inefficient for external defibrillation. Thus, our group 2 experiments did not include this capacitor.

**Clinical Implications**

To facilitate widespread dissemination and public use of such external defibrillators, it is important that these devices be reliable and appropriately priced. Thus, technological approaches that would minimize the cost of construction may be an important consideration in public access defibrillators.

Full-tilt waveforms have several potential advantages over traditional biphasic waveform designs. These advantages include higher reliability and simpler design. All of this translates into a potentially better defibrillator. In a single-capacitor biphasic waveform, a switching system is necessary to change polarity at a high voltage of phase reversal and for truncation of phase 2. Conventional biphasic waveforms have always been generated with an H-bridge-style construction. This method used electronic switches to reverse the capacitor during the middle of the discharge cycle. Such a reversal requires switches that are capable of controlling both high currents and high voltages. Switching these high currents necessitates the use of isolated-gate, bipolar transistors (IGBTs), which are fairly bulky and relatively expensive. Because the polarity of the capacitor must be totally reversed, 4 IGBTs are required. All 4 must be carefully sequenced to properly reverse the voltage. In addition, dump resistors that are frequently used to dispose of residual charges on a capacitor after truncation would not be necessary with a nearly full-tilt waveform.

A 2-capacitor construction to implement the waveforms reported here would simplify the switching requirements, because the polarity of the capacitors does not need to be reversed. Two separate capacitor banks would be used for the 2 phases. This waveform design uses the same leading-edge voltage in both phases, which allows the use of a single-charge transformer. Because virtually all of the energy is delivered to the patient, truncation circuitry is greatly simplified. This simplification means fewer parts and less energy switching, which improves the reliability of the system while reducing space and power requirements.

**Conclusions**

The major findings of this external defibrillation study are as follows. (1) For biphasic waveforms using 2 separate and almost completely discharging (95% tilt) capacitors, the phase-2 capacitor size is an important factor in maximizing defibrillation efficacy. (2) Biphasic waveforms using 2 separate and fully discharging capacitors appear to function best with a phase-2 capacitor \( \approx 20 \) to 40 \( \mu F \) for phase-1 capacitors in the 30- to 120-\( \mu F \) range. (3) In an external defibrillator for human use, the 60/20- or 60/30-\( \mu F \) may be the optimal choice among the waveforms tested here, because the 120-\( \mu F \) capacitor may generate exceedingly long pulse widths and the 30-\( \mu F \) model would use higher voltages, possibly generating greater costs without additional benefits. (4) The use of simpler components in a high-tilt biphasic waveform as described here may improve the reliability and the expense of manufacturing external defibrillators.
Acknowledgments
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References
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