Relationship between Canine Transthoracic Impedance and Defibrillation Threshold

Evidence for Current-based Defibrillation

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Abstract

The electrical parameter used to define defibrillation strength is energy. Peak current, however, may more accurately reflect the field quantities (i.e., electric field strength and current density) that mediate defibrillation and therefore should be a better clinical descriptor of threshold than energy. Though transthoracic impedance is a major determinant of energy-based threshold and is sensitive to operator-dependent changes in impedance (electrode-subject interface), an ideal threshold descriptor should be invariant with respect to these changes in impedance. We therefore compared the relative invariance of energy- and current-based thresholds when transthoracic impedance was altered by one of two methods: (*a*) change in electrode size (protocol A) or (*b*) change in electrode force (protocol B). In protocol A, impedance was altered in each dog by a mean of 95%. Energy thresholds determined at both low and high impedance were 44±21 J (mean±SD) and 105±35 J, respectively, $P < 0.0001$. In contrast, peak current (A) thresholds were independent of transthoracic impedance, 22±5 A (low impedance) vs. 24±6 A (high impedance), P = NS. Energy and current thresholds showed a similar relationship for animals tested in protocol B. Therefore, current-based thresholds, in contrast to energy thresholds are independent of operator-dependent variables of transthoracic impedance and are invariant for a given animal. These results suggest that redefining defibrillation threshold in terms of peak current rather than energy provides a superior method of defibrillation.

Introduction

Considerable controversy exists regarding the electrical parameter that best describes defibrillation threshold. The standard clinical practice is to deliver a shock calibrated in units of energy. However, field quantities such as myocardial electric field strength and current density actually describe the electrical parameters of defibrillation (1). Peak current may more precisely reflect these field quantities than energy (for a damped sine wave pulse). Identification of the appropriate threshold descriptor has significant clinical importance since a suprathreshold shock can cause MB-creatine kinase release (2), frank myocardial necrosis (3, 4), and give rise to shock-induced arrhythmias that may immediately refibrillate the heart (5, 6).

Clinical studies have yielded conflicting data regarding the optimal energy dose required to defibrillate a patient (7–10). Despite significant subject-to-subject variability in energy thresholds (unrelated to body weight (11–14), the consensus recommendation is to deliver a fixed energy dose to all patients (15). This energy-based method of defibrillation may however, result in delivering excessive current to patients with low transthoracic impedance and insufficient current to patients with high impedance.

To date, no study has shown that peak current (A) is a better descriptor of threshold than energy. Rather, experimental studies have only shown that peak current normalized to body weight (1 A/kg) (16) is a better threshold descriptor than energy (17). These studies have had minimal clinical impact since delivering 1 A/kg would result in markedly excessive current doses compared with that delivered by present energy-based practice.

Transthoracic impedance is a major determinant of energy-based defibrillation threshold (14, 18, 19). Transthoracic impedance is not, however, constant for a given subject, but rather is sensitive to and dependent on the external thoracic interface between the electrodes and subject. We hypothesized that these operator-dependent variables of impedance (e.g., electrode force, size, and electrode-electrolyte interface) primarily affect series rather than parallel changes in the electrode-subject interface. Therefore, current-based thresholds for a given subject should be invariant and independent of changes in transthoracic impedance in contrast to energy- and voltage-based thresholds. We thus examined the relationship between variations in transthoracic impedance and defibrillation threshold as defined by delivered energy, peak voltage and peak current.

Methods

General

Mongrel dogs weighing 16 to 24 kg were anesthetized with pentobarbital, 30 mg/kg, and ventilated with a Harvard respirator. Electrocardiographic surface lead II and arterial pressure (Statham P23Db transducer, Statham Instruments, Inc., Oxnard, CA) were continuously monitored on an electrostatic recorder (model ES-1000, Gould, Inc., Houston, TX). Electrodes covered with electrode paste (Redux Paste, Hewlett-Packard, Inc., Palo Alto, CA) were firmly placed over the shaven right and left lateral chest walls at the transverse level of the heart. Damped-sinusoidal DC shocks were delivered by either an unmodified, commercially available defibrillator (Lifepak 6, Physio-Control Corp., Redmond, WA) in which the energy setting is equivalent to the energy discharged into a 50 Ω load (protocol A) or by a modified device in protocol B.

Voltage delivered across the thorax was measured with a 1,000:1 voltage divider in parallel with the defibrillator output, and the delivered current was measured with a 0.10 Ω resistor in series with the

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defibrillator output. Voltage and current waveforms were displayed on a triggered-sweep storage oscilloscope (model 5113, Tektronix, Beaverton, OR) with a frequency response from DC to 1 MHz.

Time course of voltage and current during discharge

To derive a simplified relationship between peak current, energy and transthoracic load, the time courses of delivered voltage and current waveforms were examined across a range of six energy settings (30 to 400 J) in 13 dogs as well as between all dogs at 400 J. The stored waveforms were photographed with a Polaroid camera and then traced onto the digitizing tablet of a microcomputer (Hewlett-Packard 9815) for analysis.

Relationship between transthoracic impedance and energy-, voltage-, and current-based defibrillation thresholds

To determine the relationship between transthoracic impedance and energy-, voltage- and current-based thresholds two protocols were designed.

Protocol A (alteration of impedance by electrode size). Defibrillation thresholds were determined in 11 mongrel dogs at two different impedances (high and low) using a standard defibrillator. Transthoracic impedance was altered by a mean of 95% (range, 50 to 188%) in each animal by using different sized oval electrodes (19 cm^2-high impedance or 76 cm^2-low impedance). Ventricular fibrillation was induced by introducing alternating current through a percutaneous quadripolar catheter (No. 6F, United States Catheter and Instrument Corp., Billerica, MA) positioned either in the right or left ventricle. After 30 s of ventricular fibrillation, oval electrodes covered with electrode paste were firmly held by the operator over the right and left lateral chest wall at the approximate transverse level of the heart. Energy was discharged at an initial setting of 30 J during peak inspiration. Incremental energy settings were selected (50, 100, 200 J) until defibrillation was achieved. Peak current and energy were recorded for each shock and the transthoracic impedance calculated.

To confirm reproducibility of threshold measurements (at each impedance level) in each dog, ventricular fibrillation was reinitiated during three additional trials following a 5-min equilibration period during which blood pressure, arterial blood gases, and heart rate returned to baseline. Results were accepted for analysis if thresholds for at least three of four trials were identical at each of the two impedance levels. Defibrillation thresholds were determined initially at low impedance and then repeated at high impedance.

Protocol B (alteration of impedance by electrode force). Transthoracic impedance was altered in this protocol by changing electrode force through a precision force-control system. Thresholds were determined at high electrode force, 130 Newtons (N) (low impedance) and at low electrode force, 20 N (high impedance). A defibrillator modified specifically for this protocol (Fig. 1) was calibrated in units of current and discharged through a precision control system that regulated electrode force. Electrode size (60 cm^2, circular stainless steel) was held constant.

The current-based defibrillator (constant-load current divider circuit) was a modified commercially available defibrillator that was calibrated to deliver 3,000 V at 400 J across a 50 Ω load. The dog's transthoracic impedance or resistance R_T was determined prospectively with a 30-kHz sine wave (18, 20) transmitted through the external electrodes. The variable resistors R_P and R_S (Fig. 1) were then adjusted to deliver the desired current to R_T and simultaneously maintain a constant 50 Ω load to the defibrillator. Thus, a slightly overdamped pulse was delivered (that was within 2% of the selected current) during each defibrillation trial, regardless of the transthoracic impedance.

The system to control electrode force was adjusted by changing precision weights on a pair of levers. The tension of the weights was changed to a compressive force on the electrodes by a statically-balanced pair of levers, each having one horizontal and one diagonal arm (Fig. 2). All forces due to the weight of the levers and electrodes were

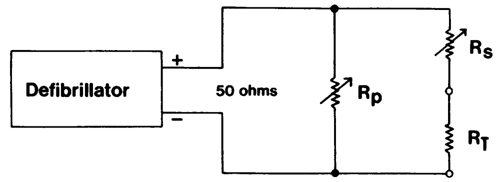

Figure 1. Modified current-based defibrillator (constant-load current divider circuit). See text for details. R_P, variable parallel resistor; R_S, variable series resistor; R_T, transthoracic resistance (subject).

cancelled by adjusting counter-weights until the horizontal beams were statically balanced with no applied weight. Thus, the net force at the electrodes was equal in magnitude to that supplied by the precision weights. The endotracheal tube was clamped at peak-inspiriation before balancing the system and delivering the shock.

Defibrillation thresholds were determined at both high and low electrode force. After induction of ventricular fibrillation, thresholds were determined, starting with 12 A and incrementing in steps of 2 A until defibrillation occurred. Thresholds were determined in three consecutive trials with low electrode force (high transthoracic impedance) at 5-min intervals. Then three consecutive thresholds were determined with high electrode force (low transthoracic impedance) at 5-min intervals. Preliminary experiments performed in four dogs to determine the stability of current-based thresholds with respect to number of shocks and time showed an approximate exponential decline of 30% in threshold over the first six shocks (5-min intervals), which was followed by a stable plateau phase. For this reason, thresholds during the experimental protocol were determined after six shocks were delivered in sinus rhythm at 5-min intervals. Voltage, current, and energy thresholds and transthoracic impedance determined during the plateau phase were then averaged for all defibrillation trials in each electrode configuration.

Statistical analysis

Statistical comparisons of energy, voltage and current defibrillation thresholds at two different transthoracic impedances in each animal were performed using Student's paired t test. The significance of linear relationships between variables was examined using least squares linear regression. Data are expressed as mean±SD. Differences were considered significant for $P < 0.05$.

Results

Relationship between transthoracic impedance and defibrillation threshold

Protocol A: alteration of impedance due to electrode size. Defibrillation thresholds determined in each animal at a control impedance, 44±8 Ω (low impedance, standard electrodes) and at an impedance increased by 95%, 86±17 Ω (small electrodes) (range, 50 to 188%) showed marked variability with respect to energy and voltage thresholds. For example, at low impedance threshold energy was 44±21 J compared with 105±35 J at high impedance, $P < 0.0001$. Threshold voltage was 937±237 V at low impedance compared with 1,980±291 V at high impedance, $P < 0.0001$ (Fig. 3). In contrast, peak current thresholds remained invariant with respect to impedance, 22±5 A (low impedance) vs. 24±6 A (high impedance), P = NS. Furthermore, current showed a smaller coefficient of variation (SD/mean) than energy, 24 vs. 41%. Thus, each animal demonstrated a threshold current that was independent of transthoracic impedance, whereas energy and voltage requirements were directly related to impedance for a given animal.

Protocol B: alteration of impedance due to electrode force. Results were similar to those observed in protocol A. Low

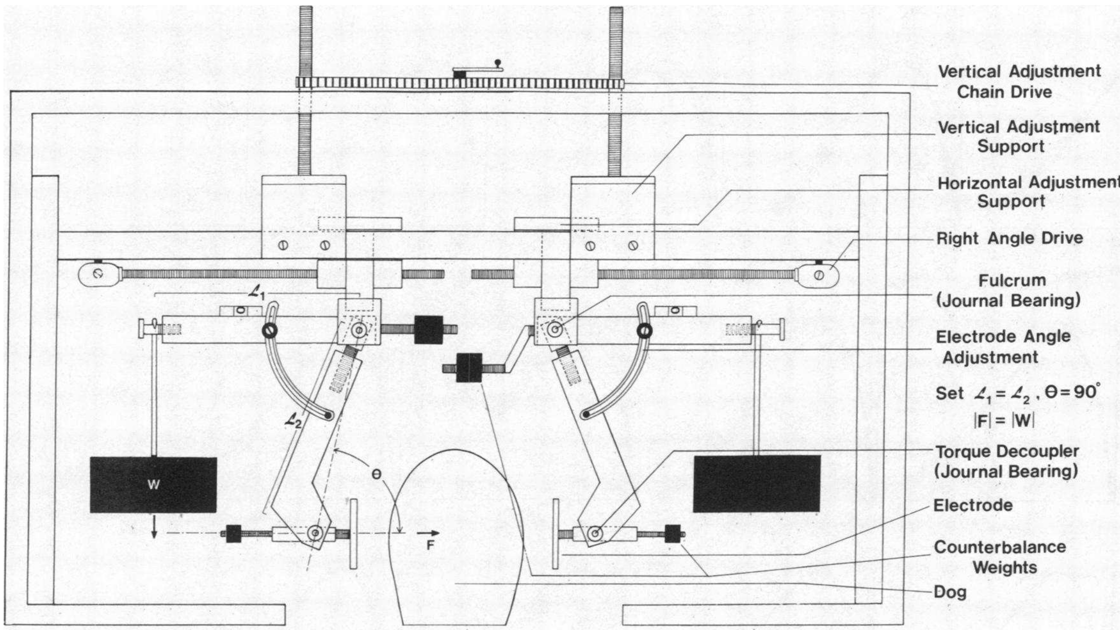

Figure 2. Apparatus for controlling electrode force. See text for discussion.

electrode force (20 N) increased the transthoracic impedance in each animal by a mean of 32%, from 71±9 Ω at high electrode force (130 N) to 94±10 Ω at low electrode force, $P < 0.0001$. Energy thresholds increased in each dog with low electrode force, from 70±34 J (130 N) to 111 ± 47 J(20 N), $P < 0.001$ as did voltage thresholds, 1,452 ± 363 V (130 N) to 2,117±489 V (20 N), $P = 0.0001$ (Fig. 4). In contrast, current thresholds remained invariant with respect to change in transthoracic impedance, 21±6 A (low impedance) and 23±5 A (high impedance), P = NS. In addition, the coefficient of variation was less for current than for energy-based thresholds, 25 vs 45%. An initial shock of 30 A would have defibrillated 90% of the animals in protocols A and B. Current thresholds were not related to body weight in either protocol A or B.

Relationship between peak current, energy, and transthoracic impedance

To determine whether the time courses of voltage and current waveforms were dependent on the magnitude of the discharged pulse, superimposition of the normalized voltage and current waveforms was performed in each dog at the six energy levels (Figs. 5 and 6). Similarly, the data were examined for inter-animal variability by normalizing the 400-J shocks across all dogs (Fig. 7). The curves in these figures show that the time courses of delivered voltage and current were similar across the range of shocks in each dog as well as between all dogs at a constant energy level.

Since the curves all have a similar qualitative shape and the transthoracic impedance is predominantly resistive (21, 22), a

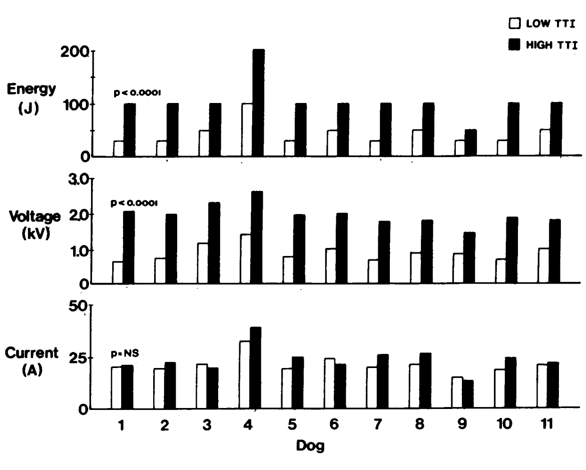

Figure 3. Energy-, voltage- and current-based thresholds as a function of transthoracic impedance. Impedance was changed in each animal by a mean of 95% by altering electrode size. TTI, transthoracic impedance. See text for discussion.

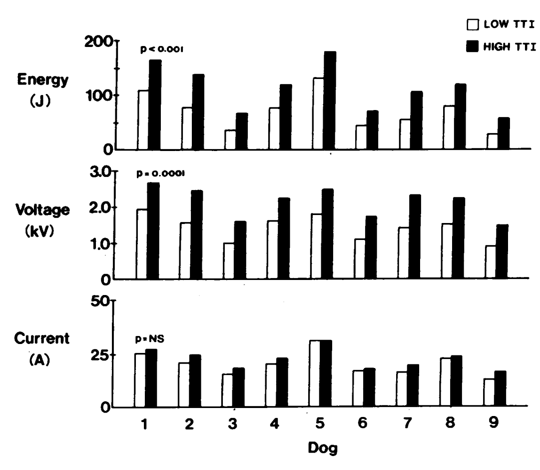

Figure 4. Energy-, voltage- and current-based thresholds. Impedance was changed in each animal by altering electrode force. Abbreviations as previously designated. See text for discussion.

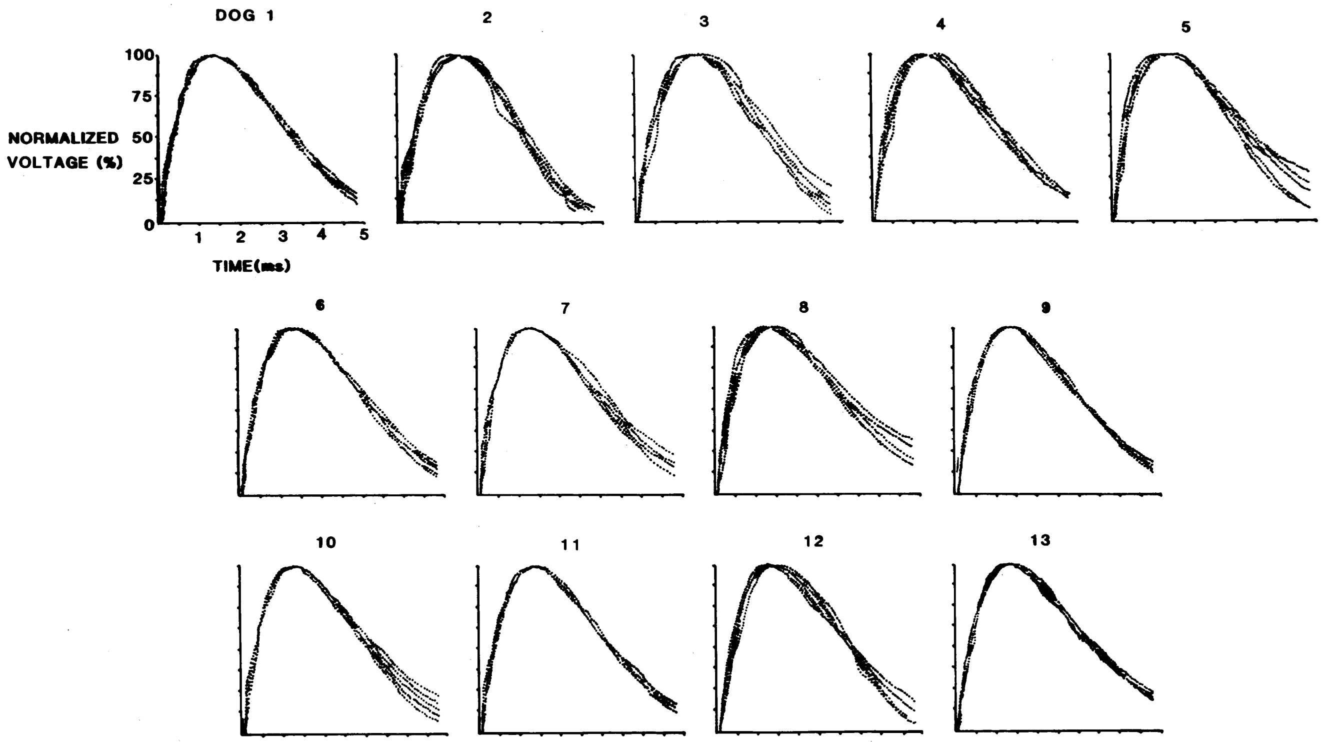

Figure 5. Each panel displays the voltage waveforms for each dog at 6 energy levels, (30–400 J). Each waveform was normalized to a peak value of 100. The time course of voltage decay is independent of the magnitude of discharged energy.

relationship between peak delivered current and discharged energy can be derived in the following manner: By definition,

$$W = \int i^2(t)Rdt \qquad (1)$$

where W is delivered energy, $i(t)$ is the value of current at time t and R is transthoracic resistance. By dividing current by its peak value, I_p, one can obtain a function for normalized current $i_N(t)$:

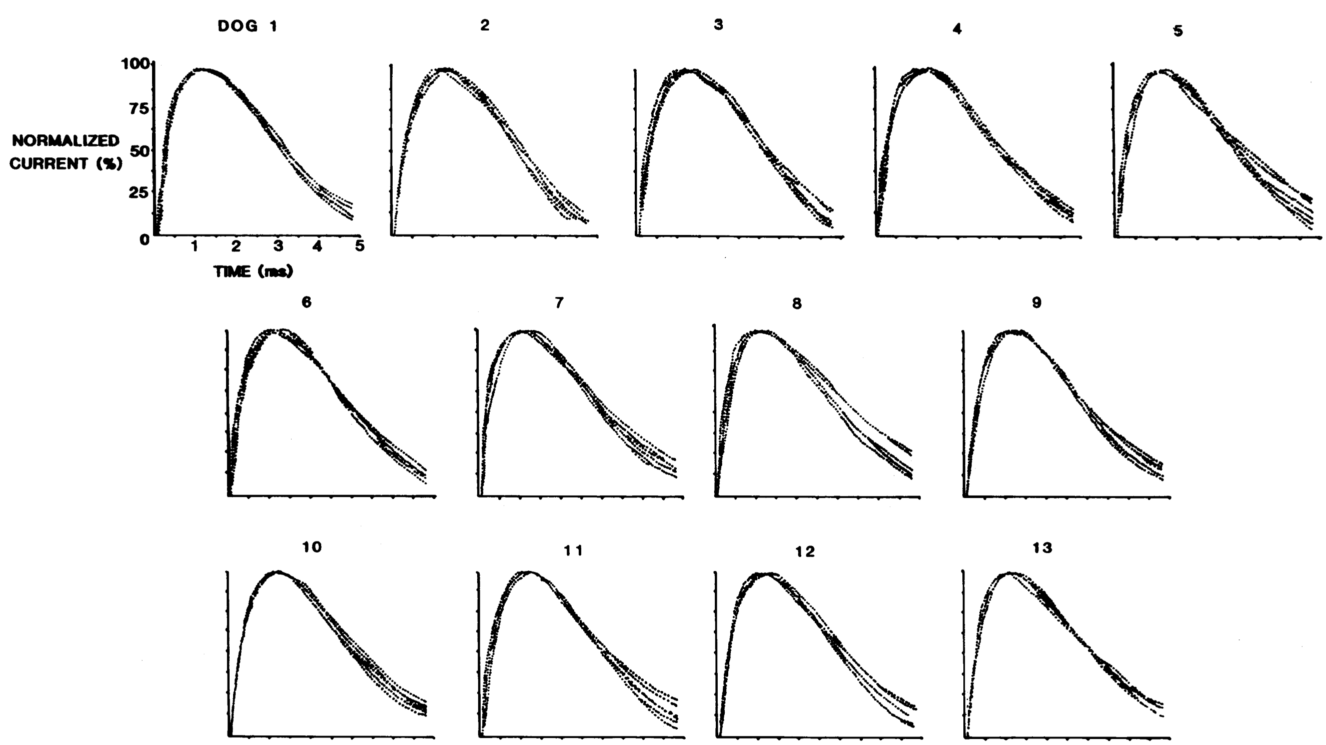

Figure 6. Normalized current waveforms for each dog during 6 different energy pulses (30–400 J). The time course of delivered current is independent of the magnitude of energy and the time courses of voltage and current are identical to each other.

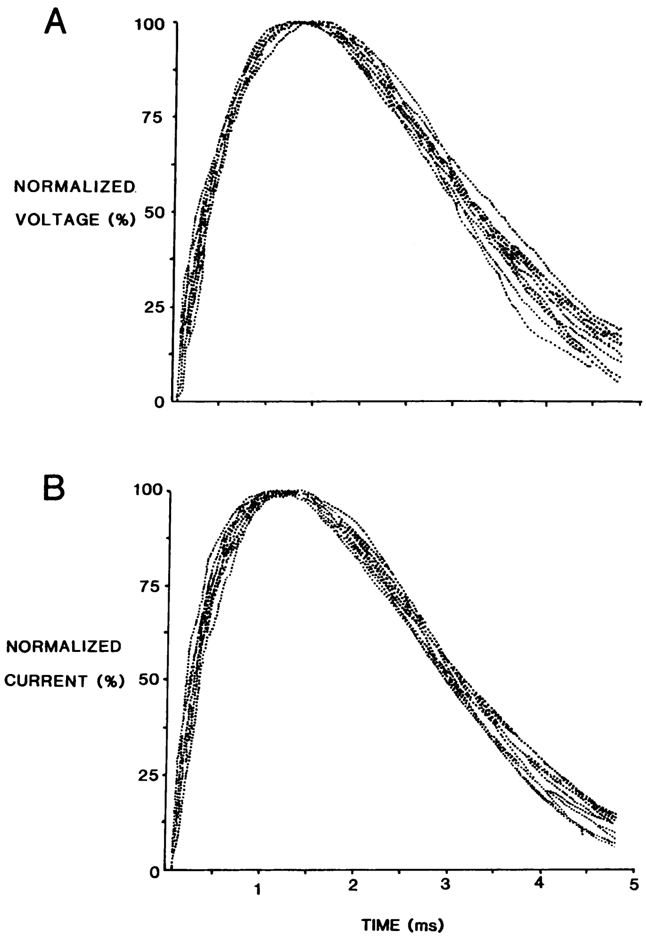

Figure 7. (*A*) Normalized voltage waveforms superimposed for 13 dogs during a 400-J shock. Each waveform was normalized to a peak of 100. The time course of voltage decay was nearly identical for all dogs. (*B*) Normalized current waveforms superimposed for 400 J shocks. The time course of current decay was nearly identical for all dogs and therefore was independent of the transthoracic load.

$$i_N(t) = i(t)/I_p \tag{2}$$

Substituting Eq. (2) into Eq. (1),

$$W = \int I_P^2\, i_N^2(t) R dt = I_P^2 R \int i_N^2(t) dt \tag{3}$$

The shapes of the normalized current waveforms across each dog and between all dogs were nearly identical and superimposable for the range of energies examined (30–400 J) (Figs. 6 and 7). Under these conditions, $\int i_N(t)dt$ is constant as is

$$\int i_N^2(t)dt.$$

Setting this latter integral to K_1 we may derive

$$W = I_P^2 R K_1 \tag{4}$$

and solving for I_p, we have an equation relating peak current to delivered energy and transthoracic resistance:

$$I_p = K_2 \sqrt{W/R} \tag{5}$$

where $K_2 = 1/\sqrt{K_1}$.

The data derived from discharging six energy levels across 13 dogs were used to evaluate the validity of Eq. 5 according to a general linear model (Statistical Analysis System, SAS Institute, Cary, NC). This procedure permitted the assessment of the overall slope and intercept of the relationship and the likelihood that any individual animal had a different slope from the others (homogeneity of slopes). Differences in slope between animals were not significant (P = NS), and thus an overall slope and intercept were calculated from all animals in a combined model. The data were strongly linearly related ($r = 0.999$, $P < 0.0001$) with the regression equation:

$$I_p = 22 \sqrt{W/R} - 0.9 \tag{6}$$

The regression line was then force-fitted through zero with the resultant equation (Fig. 8) ($r = 0.999$, $P < 0.0001$):

$$I_p = 22 \sqrt{W/R} \tag{7}$$

Predictive accuracy of derived current equation

To prospectively validate the predictive accuracy of Eq. 7, six shocks at energy levels from 30 to 400 J were discharged transthoracically in an additional 10 dogs. The energy setting on the defibrillator was used as an approximation for delivered energy since the two are nearly equivalent for the range of transthoracic impedance observed in this study.. The estimates for peak current derived from Eq. 7 are presented together with the measured values for peak current in Fig. 9. For the six energy levels across the 10 dogs, 92% of the estimates derived from Eq. 7 were within 5% of the measured values, and 100% of the estimates were within 8% of the measured values. There was also no significant difference (ANOVA) between measured peak current, predicted peak current by Eq. 7, and that predicted by second order source-free RLC circuit equations (23), which completely describe the defibrillator circuit and load (Fig. 9).

Discussion

The major findings of this study are that current-based thresholds, in contrast to energy- and voltage-based thresholds, are invariant for a given subject and are independent of operator-dependent variables of transthoracic impedance. Impedance was altered in each animal by either varying electrode size or force. These methods of altering impedance were chosen since changes in the electrode-subject interface are operator-depen-

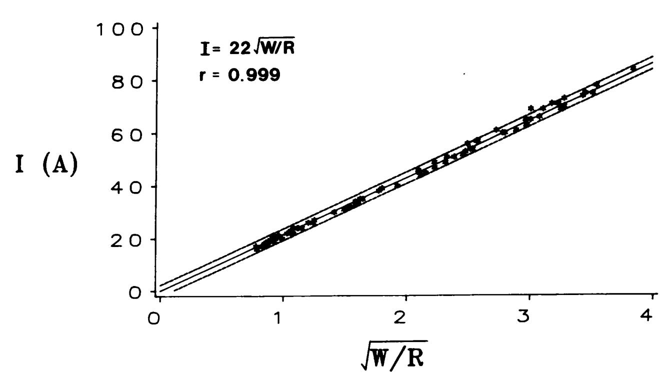

Figure 8. Linear regression for peak delivered current (I) as a function of the square root of the quotient of delivered energy (W) and transthoracic load (R).

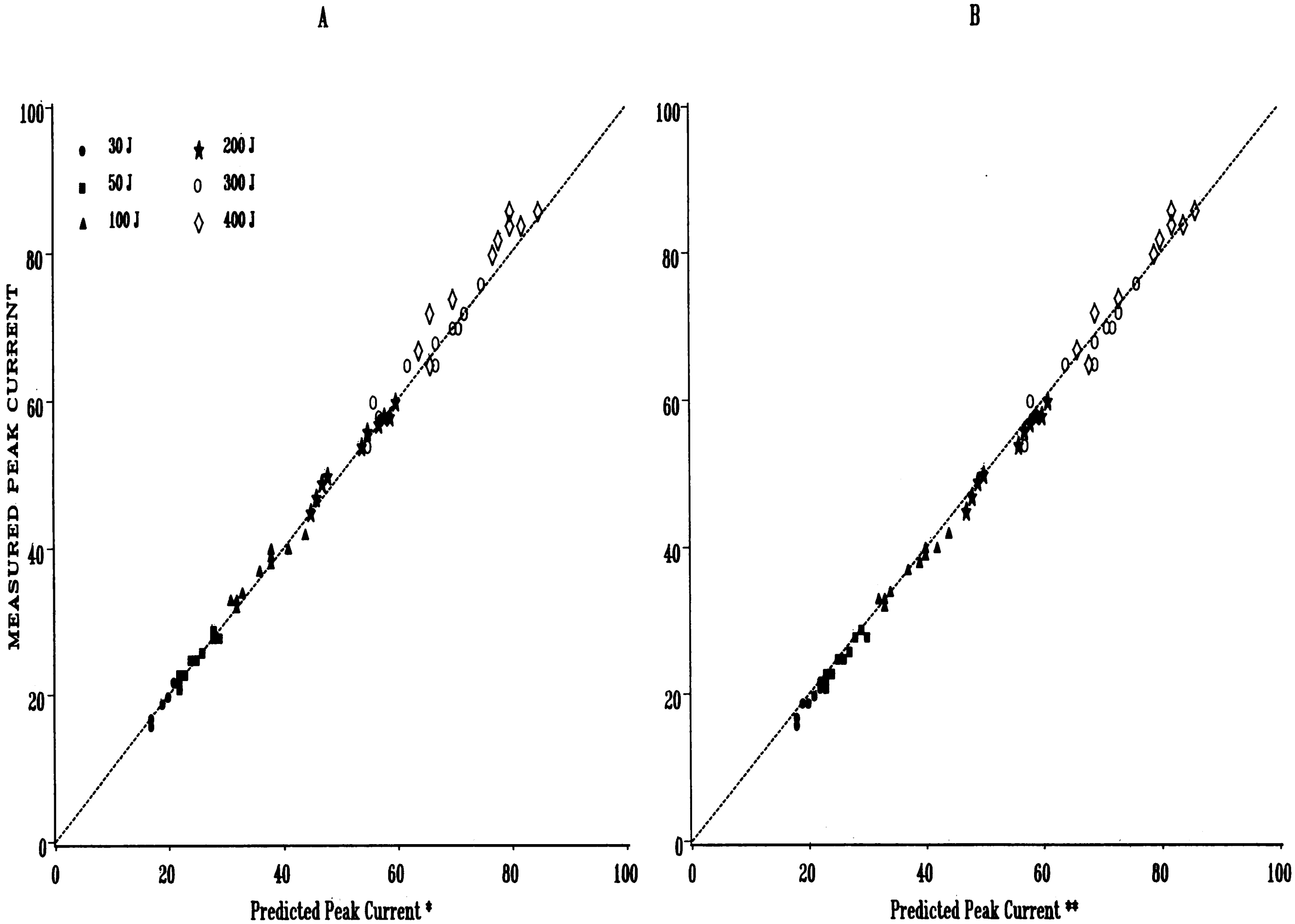

Figure 9. Comparison between measured and predicted peak current across 10 animals at energy levels 30 to 400 J. (*A*) Predicted peak current determined by Eq. (7)*. Dotted line represents the line of identity. (*B*). Predicted peak current based on second order source-free RLC equations (23)**. Capacitance, 36 μF; inductance, 28 mH; internal resistance of defibrillator, 12.8 Ω.

dent and can alter significantly between successive shocks (24, 25).

An ideal descriptor of threshold should be invariant with respect to operator-dependent changes in impedance. However, these changes in impedance are a major determinant of defibrillation success in the energy-based approach since high impedance patients require more energy to defibrillate than low impedance patients (14, 18, 19). Experimentally, impedance-based energy adjustments have been shown to improve defibrillation rates when impedance is increased by altering the electrode-subject interface (19). Our data indicate, however, that this approach is indirect and suboptimal since (*a*) this method dictates that both energy and current doses are dependent on transthoracic impedance, and (*b*) in contrast to energy thresholds, current thresholds are invariant and insensitive to changes in transthoracic impedance.

An additional limitation of the energy-based defibrillation method is the difference between indicated (or selected) energy on a defibrillator and the actual delivered energy. For a given indicated energy, actual delivered energy will show significant variability across all subjects due to its dependence on transthoracic impedance. This concept can be understood by examining the relationship between three discrete but commonly misunderstood variables: W_s, stored energy (capacitor); W_d, actual delivered energy; and W_i, indicated or selected energy on a defibrillator. For example,

$$W_d \cong W_s\left(\frac{R_T}{R_T + R_0}\right) \tag{8}$$

where R_T = transthoracic resistance, and R_0 = internal resistance of the defibrillator. For a 50-Ω load, delivered energy is equivalent to indicated energy (W_i). That is,

$$W_i \cong W_s\left(\frac{50\Omega}{50\Omega + R_0}\right) \tag{9}$$

W_i can be expressed in terms of stored energy W_s by rearranging Eq. 9.

$$W_s \cong W_i\left(1 + \frac{R_0}{50\Omega}\right) \tag{10}$$

Substituting Eq. 10 into Eq. 8 gives the relationship between indicated and delivered energy

$$W_d \cong W_i\left(1 + \frac{R_0}{50\Omega}\right)\left(\frac{R_T}{R_t + R_0}\right) \tag{11}$$

Therefore, according to Eq. 11, a shock discharged at a 200-J setting will deliver more energy to a 100-Ω subject than to a 50-Ω subject.

Though the concept of defibrillation defined in terms of current has been previously introduced (16), the dose was considered weight dependent and was therefore normalized to body weight (1 A/kg). Minimal interest in this approach developed since it would deliver 70 A to the average person and more than 100 A to large subjects, a dose far in excess of that delivered by present energy-based practice (26).

Our data do not address the threshold-impedance relationship due to changes in resistance of intrathoracic structures. These changes, though potentially important, develop more slowly and are less frequently encountered clinically during successive shocks than are electrode-subject alterations in impedance. One would predict that the development of pleural effusion, pleural fibrosis, pulmonary edema or pericardial effusion, etc., could potentially affect changes in resistances both in series and parallel to the transthoracic electrodes. Under some of these conditions, therefore, voltage rather than current could prove to be a better threshold descriptor.

The electrical strength for defibrillation pulses has customarily been calibrated in units of energy. It is likely that energy-based defibrillators were originally developed because of technological considerations, one of which was the ability to measure the energy stored in a capacitor. The simple relationship derived in this study between transthoracic resistance (impedance), energy and peak current (Eq. 7) provides a clinically useful relationship that permits the operator to select the energy setting necessary to deliver a selected peak current using a standard, commercially available defibrillator (provided that transthoracic impedance is prospectively determined). Automated current-based defibrillators can be developed for clinical use with microprocessor-based technology.

Though our data show variability in current-based thresholds across all dogs, it is considerably less than that for energy or voltage. Our results suggest that a fixed-dose of 30 A will defibrillate most animals. Extrapolation of published clinical data, though not analyzed or interpreted in this respect, suggest that an initial shock between 20 and 30 A would be an appropriate first approximation for human defibrillation (11, 18).

In summary, our findings demonstrate that current-based thresholds, in contrast to energy and voltage thresholds, are independent of operator-dependent variables of transthoracic impedance and are invariant for a given subject. These results suggest that redefining defibrillation threshold in terms of peak current rather than energy provides a superior method of defibrillation.

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