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The transthoracic impedance cardiogram is a potential haemodynamic sensor for an automated external defibrillator

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Aims The American Heart Association has endorsed the concept of Public Access Defibrillation. However, there have been reports of inappropriate direct current shocks from automatic external defibrillators. The specificity of automatic external defibrillators for shockable rhythms may be improved by the incorporation of a haemodynamic sensor.

Methods and Results This study examined the use of four parameters extracted from the impedance cardiogram i.e. Peak dz/dt (the peak of the impedance cardiogram measured from the line dz/dt=0 Ωs^{-1}), Peak-trough (the peak-to-trough measurement of the impedance cardiogram Ωs^{-1}), Area 1 (the area under the C wave of the impedance cardiogram above the line dz/dt=0 mΩ) and Area 2 (the area under the impedance cardiogram 50 ms on either side of the Peak and above the line dz/dt=0 mΩ) as predictors of cardiac output.

At 116 cardiac arrest calls the ECG and impedance cardiogram were recorded through two ECG/defibrillator pads placed in an antero-apical position. Nine recordings were rejected for artefact. The rhythm recorded in the remaining 107 calls was asystole (19), ventricular fibrillation (14), agonal rhythm (20), electromechanical dissociation (22), ventricular tachycardia (27) and sinus rhythm (5). These rhythms were divided into those associated with haemodynamic collapse i.e. no pulse — asystole, ventricular fibrillation, agonal rhythm, electromechanical dissociation and shockable ventricular tachycardia (associated with loss of consciousness, pulselessness or a systolic blood pressure of <80 mmHg) (Group 1) and those associated with a satisfactory cardiac output i.e. non-shockable ventricular tachycardia (conscious with a pulse) and sinus rhythm (Group 2). On univariate analysis each of the four impedance cardiogram parameters were significantly greater in Group 2 than Group 1 (P<0·001). On multivariate analysis the parameters which best differentiated the two groups were Area 1 and Peak-trough.

Conclusion Thus the impedance cardiogram is a potential haemodynamic sensor for an automatic external defibrillator.
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Key Words: Impedance cardiogram, automatic external defibrillator, cardiac arrest.

Introduction

Survival from out-of-hospital ventricular fibrillation has been possible since the development of the portable defibrillator[1]. The most important factor influencing survival is the delay to defibrillation[2]. The scope of early defibrillation has been extended by the development of the automatic external defibrillator. This device has been used successfully by family members of survivors from sudden cardiac death and by lay persons in large public gatherings[3,4]. The American Heart Association has endorsed the concept of Public Access Defibrillation[5]. However, there have already been reports of the inappropriate delivery of direct current shocks to patients without ventricular tachyarrhythmias[6–8]. This has led the U.S. Food and Drug Administration to issue a safety alert with one of these devices[8]. The use of automatic external defibrillators by minimally trained individuals who lack the skills to differentiate cardiac arrest from other causes of collapse will increase the potential for inappropriate defibrillation. Clearly, the inclusion of a non-ECG haemodynamic sensor in an automatic external defibrillator device could increase its specificity.

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Impedance cardiography is a non-invasive method for measuring cardiac output, and parameters derived from the impedance cardiogram have been used as indices of myocardial contractility and aortic blood flow[9–18]. The impedance of the thorax (Z) can be recorded by passing a high frequency low amplitude current between two electrodes and recording the resulting voltage. Ejection of blood into the aorta causes small fluctuations in this impedance (ΔZ). The impedance cardiogram is a recording of the first time derivative of ΔZ (i.e. dz/dt) against time[19].

Traditionally the impedance cardiogram has been recorded using four circumferential band electrodes[19]. The upper voltage electrode is placed around the base of the neck and the lower voltage electrode around the thorax at the level of the xiphisternum. The two outer current electrodes are placed at least 3 cm away from the voltage electrodes. Clearly it was not practical to employ such an electrode configuration in the cardiac arrest setting. Thus, a system was developed in which the impedance cardiogram could be recorded through two ECG/defibrillator pads, one placed at the second right intercostal space to the right of the sternum just beneath the clavicle and the other placed over the fifth left intercostal space in the mid-clavicular line.

The purposes of this research were firstly, to compare the impedance cardiogram recordings using a traditional four band electrode technique with the novel two ECG/defibrillator pad technique and secondly, to determine if one or more of the impedance cardiogram parameters, recorded through two ECG/defibrillator pads, could act as a haemodynamic sensor for an automated external defibrillator.

Methods

Controls and patients

The impedance cardiogram was recorded in 20 male subjects (mean ($\pm$SD) age, 26·1 ($\pm$5·8)) in sinus rhythm using sequentially the traditional four-band electrodes and the new two ECG/defibrillator pad technique.

Over a 14-month period, simultaneous recordings of the ECG and impedance cardiogram were made at 116 cardiac arrest calls in 110 patients. The baseline characteristics age, sex, site of arrest, delay to cardiopulmonary resuscitation and delay to intensive care were documented for each of the patients. There were 39 females and 71 males. The age of one out-of-hospital arrest patient was not known. Of the remaining 109 patients the mean age was 66 years (range 38–87 years). Sixty cardiac (51·7%) arrest calls were attended by the mobile coronary care unit outside hospital and 56 (48·3%) were inside the hospital.

Equipment

In the control subjects the ECG, impedance cardiogram and baseline impedance (Z) were recorded using a prototype device. This device could use either the four-electrode technique or the new two ECG/defibrillator pad technique. It passed a high frequency (64 kHz) low amplitude constant alternative current (1 mA RMS) between two electrodes (i.e. the outer two band electrodes in the four electrode technique and the two ECG/defibrillator pads in the two electrode technique) and recorded the impedance cardiogram through two electrodes (i.e. the inner two band electrodes in the four electrode technique or the two ECG/defibrillator pads in the two electrode technique). The signals were digitized and stored on a portable computer for subsequent analysis using a commercial software package (Snap-Master 2·00, HEM Data Corp).

Figure 1 **The ECG/impedance cardiogram unit (A) connected to a portable defibrillator (B) and two ECG/defibrillator pads (C).**

The circuitry for the two ECG/defibrillator pad technique was then incorporated into a portable ECG/impedance cardiogram recording unit and hardware was added to protect the device from high voltage direct current shocks. This unit could be connected to each of four portable defibrillators (Liteguard 6, Temtech, Bangor, N.I.) using a locking plug (Hewlett Packard) (Fig. 1). Again this unit passed a high frequency (64 kHz) low amplitude constant alternative current (1 mA RMS) between the two ECG/defibrillator pads. The ECG and impedance cardiogram were detected and the signals simultaneously digitized and stored on memory cards for analysis off-line.

Procedure

Immediately on arrival at a cardiac arrest patient, cardiopulmonary resuscitation continued or was initiated by the junior doctor manning the cardiac ambulance or attending the in-hospital cardiac arrest and ECG/defibrillator pads positioned as for cardiac arrest management. To determine the cardiac rhythm, cardiopulmonary resuscitation was momentarily discontinued and a 10 s recording of the ECG and impedance cardiogram, free of cardiopulmonary resuscitation artefact, was made.

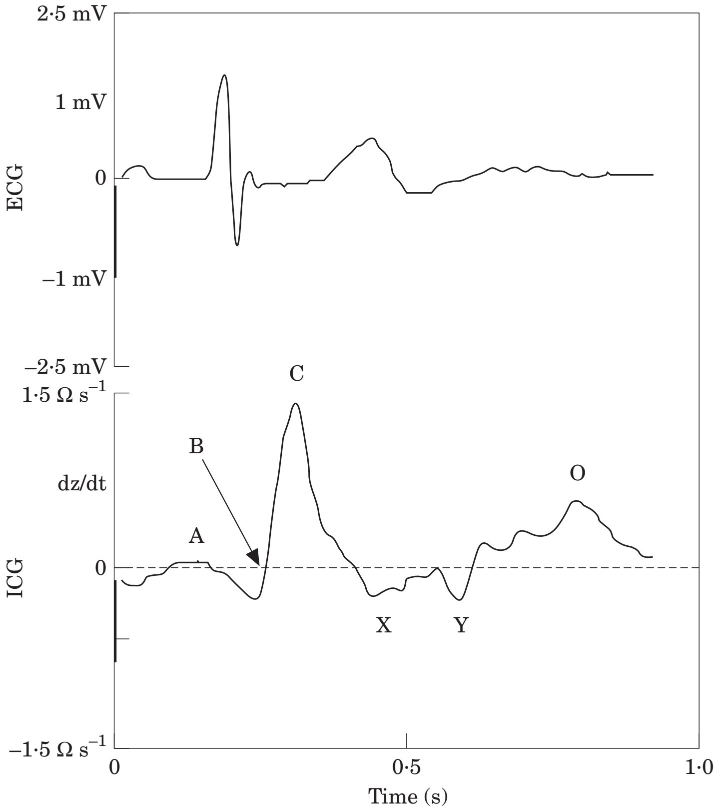


Figure 2 **Features of the impedance cardiogram on a simultaneous recording of the ECG and impedance cardiogram at 25 mms^{-1} during sinus rhythm. A=A wave representing atrial systole, B=B point corresponding with aortic valve opening, C=C wave representing ventricular systole, X=X point corresponding with aortic valve closure, Y=Y point corresponding with pulmonary valve closure and O=O wave representing ventricular filling. ICG=impedance cardiogram.**

Impedance cardiogram waveform

The impedance cardiogram waveform has A, C and O waves and contains B, X and Y points (Fig. 2). The A and C waves occur during atrial and ventricular systole, respectively, and the O wave corresponds with ventricular filling. The B point has been related to aortic valve opening and the X and Y points to aortic and pulmonary valve closure, respectively[19].

Ensemble averaging

Impedance cardiogram recordings are sensitive to motion artefact and electrical interference and thus subjected to ensemble averaging[20]. Analysis software was written by the Northern Ireland Bioengineering Centre. The peak of the R wave of the ECG was identified and used as a reference point. The ECG and impedance cardiogram signals were digitally sampled over five cardiac cycles and synchronized with the R wave. The synchronized cycles were digitally summed and averaged to provide the ensemble averaged complex. In the cases of ventricular fibrillation no R waves were present and ensemble averaging was performed using the peaks of the fibrillatory waveform that were greater than a threshold value (0·2 mV) as reference points. Similar amplitude criteria are employed in the detection algorithms of current automatic external defibrillators[21]. In agonal rhythm, the peak of the ECG complex was used as a reference point. In asystole no reference point was available and averaging was performed at second intervals over a 5 s period.

Signal analysis

Using the same analysis software, features of the ensemble averaged complex were manually extracted. These features included Peak dz/dt (the peak of the impedance cardiogram measured from the line dz/dt=0 Ωs^{-1}), Peak-trough (the peak-to-trough measurement of the impedance cardiogram Ωs^{-1}), Area 1 (the area under the C wave of the impedance cardiogram above the line dz/dt=0 mΩ) and Area 2 (the area under the impedance cardiogram 50 ms on either side of the Peak and above the line dz/dt=0 mΩ (Fig. 3(a–d)).

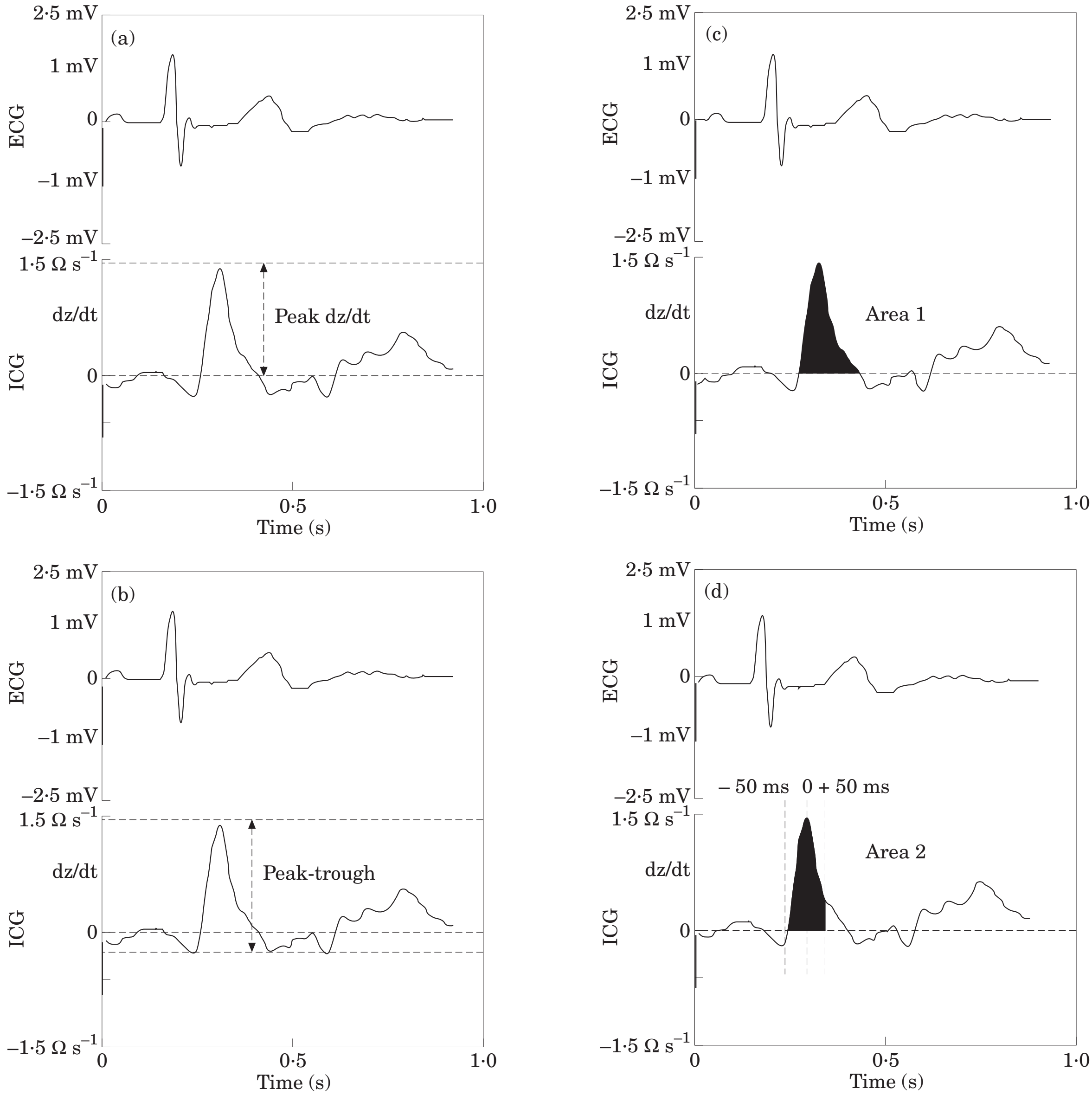


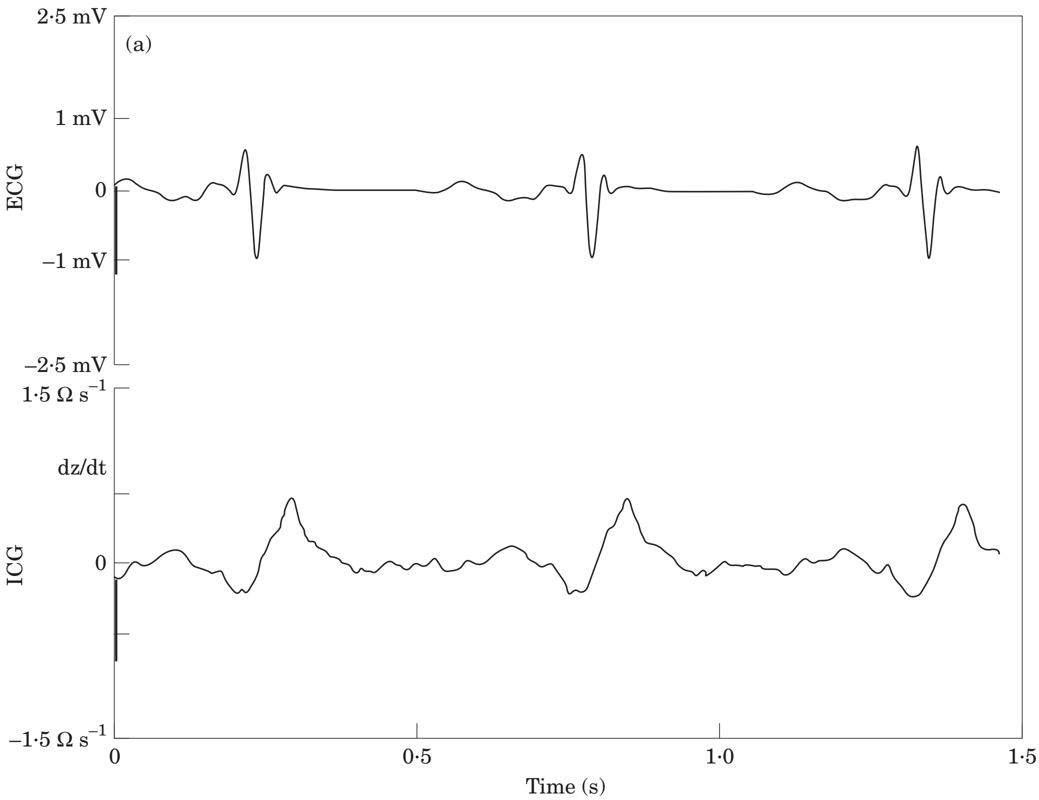
Figure 3 **The impedance cardiogram derived parameters: (a) Peak dz/dt, i.e. the maximum value of the impedance cardiogram measured from the line dz/dt=0; (b) Peak-trough, i.e. the peak-to-trough measurement of the impedance cardiogram; (c) Area 1, i.e. the area under the C wave of the impedance cardiogram above dz/dt=0; (d) Area 2, i.e. the area under the impedance cardiogram curve 50 ms on either side of the peak and above dz/dt=0.**

The analysis software also measured the R-R interval enabling a heart rate to be calculated. No R-R interval exists in either asystole or ventricular fibrillation. In agonal rhythm the interval was measured between the peaks of two consecutive complexes.

Statistics

In the control group, Peak dz/dt measured using the two-electrode technique was correlated with the four-electrode technique. Baseline impedance (Z) and Peak dz/dt in the two groups were compared using the paired t-test.

The purpose of the study was to determine if the impedance cardiogram could be used as a simple haemodynamic sensor. Thus the arrest rhythms encountered were divided into two groups: Group 1 contained rhythms associated with haemodynamic collapse i.e. no pulse — asystole, ventricular fibrillation, agonal rhythm and electromechanical dissociation or shockable ventricular tachycardia as defined by ventricular tachycardia associated with loss of consciousness, pulselessness



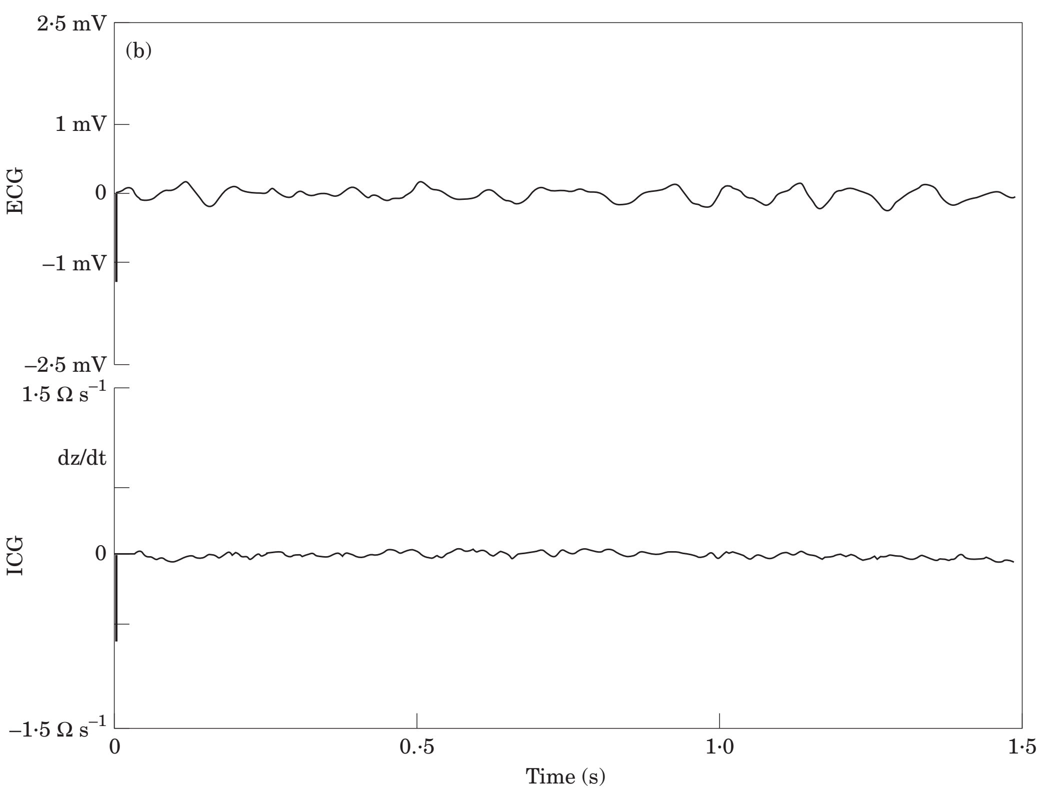
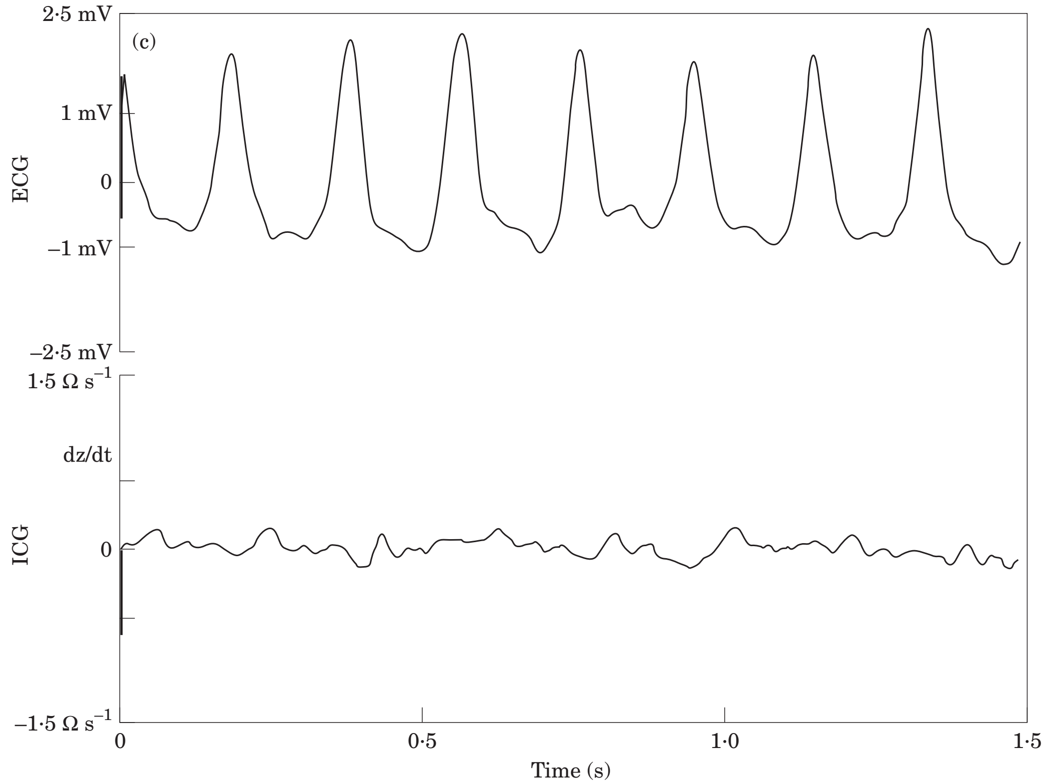


Figure 4 **(a) and (b). (See following page for legend.)**



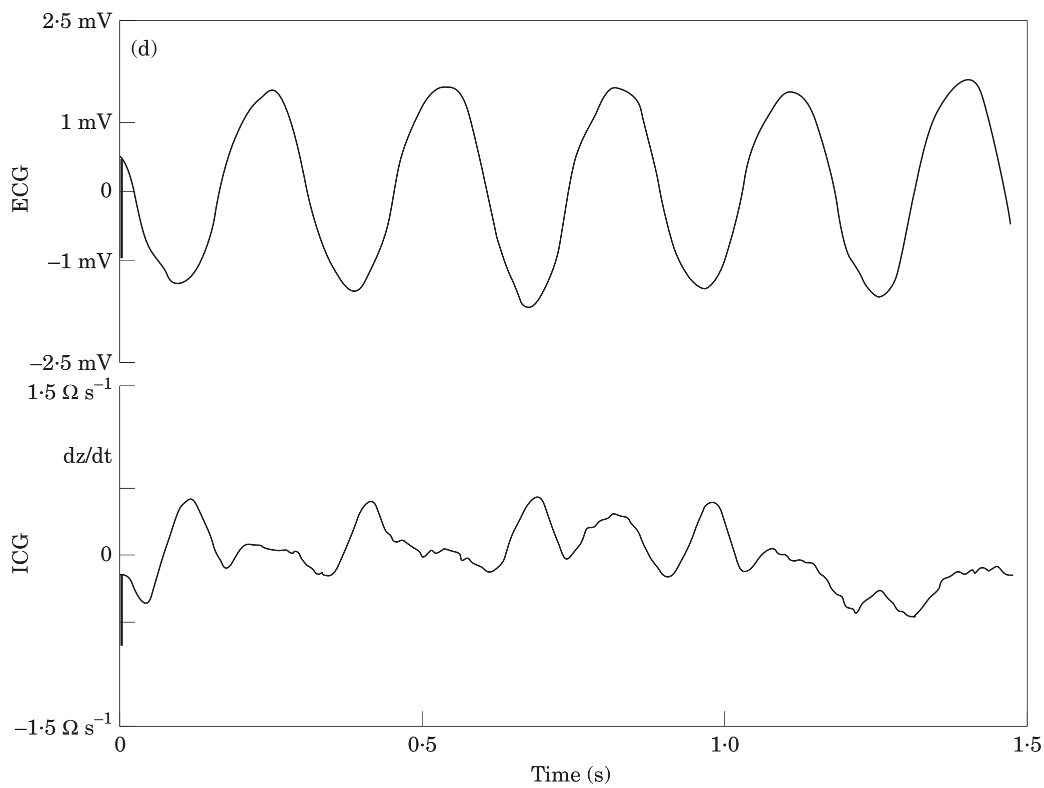


Figure 4 **(c) and (d).**

Figure 4 **The ECG and impedance cardiogram simultaneously recorded at 25 mms^{-1} during: (a) sinus rhythm, (b) ventricular fibrillation, (c) shockable ventricular tachycardia and (d) non-shockable ventricular tachycardia.**

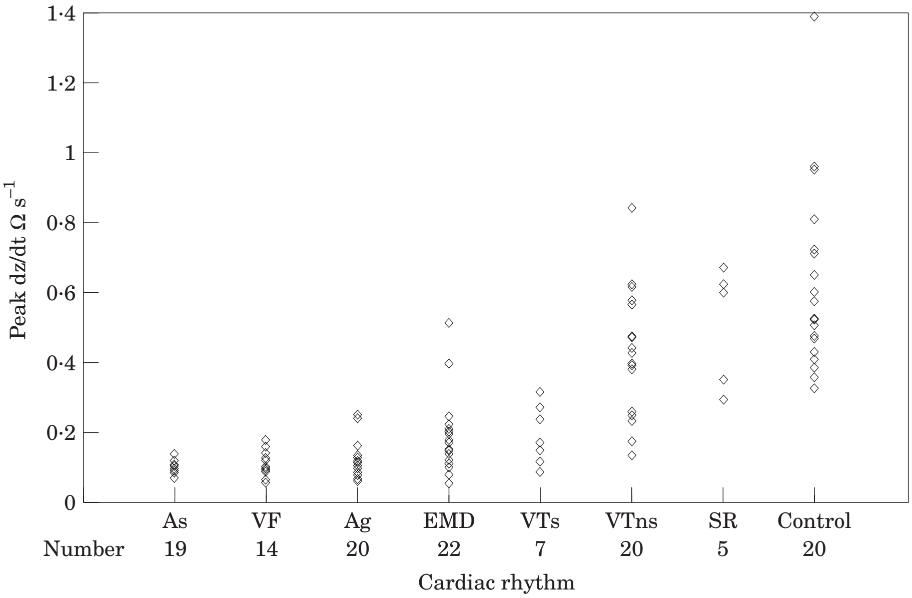


Figure 5 **Distribution of Peak dz/dt (Ωs^{-1}) by cardiac rhythm. As=asystole, VF=ventricular fibrillation, Ag=agonal rhythm, EMD=electromechanical dissociation, VTs=shockable ventricular tachycardia, VTns=non-shockable VT, SR=sinus rhythm. For comparison the 20 normal controls are included (sinus rhythm).**

Table 1 Mean (±SD) heart rate and impedance cardiogram derived parameters for each of the rhythms encountered at cardiac arrest calls

Rhythm (number)	Peak dz/dt Ωs^{-1}	Peak- Trough Ωs^{-1}	Area 1 mΩ	Area 2 mΩ	Heart rate (min^{-1})
Asystole	0·098	0·117	3·6	3·1	0
(19)	(0·017)	(0·033)	(1·3)	(1·1)	
VF	0·112	0·156	4·6	3·6	0
(14)	(0·036)	(0·071)	(2·1)	(1·5)	
Agonal	0·118	0·155	6·0	4·9	71·6
(20)	(0·052)	(0·085)	(4·5)	(3·4)	(32·2)
EMD	0·178	0·253	8·5	7·4	65·3
(22)	(0·104)	(0·119)	(6·2)	(5·9)	(27·3)
VTs	0·192	0·365	9·5	7·7	145·9
(7)	(0·085)	(0·214)	(7·9)	(6·7)	(12·7)
VTns	0·413	0·663	24·7	21·4	138·8
(20)	(0·184)	(0·312)	(12·1)	(10·1)	(31·7)
Sinus rhythm	0·505	0·738	30·5	26·3	90·5
(5)	(0·173)	(0·288)	(10·1)	(10·5)	(45·9)

EMD=electromechanical dissociation; VF=ventricular fibrillation; VTs=shockable ventricular tachycardia; VTns=non-shockable ventricular tachycardia.

or a systolic blood pressure of less than 80 mmHg (Fig. 4). Group 2 contained rhythms of patients where a cardiac arrest call was initiated but on arrival with the patient sinus rhythm was present in five (respiratory arrest) and in 20 with non-shockable ventricular tachycardia the patient was conscious with a pulse (Fig. 4).

Statistical analysis was performed using the software SPSS for Windows (SPSS UK Ltd, Surrey, England). In univariate analysis, frequency data were compared using the Chi-square test and continuous data using the t-test for independent samples. Multivariate analysis was performed using multiple logistic regression with a backstep method. Research Ethics Committee, Queen's University Belfast gave ethical approval for this study.

Results

Controls

There was a significant correlation between Peak dz/dt measured using the four-band electrode technique and

Table 2 Baseline characteristics of 107 cardiac arrest calls

	Group 1	Group 2	Significance P
Age (mean (±SD) years)	65·5 (11·5)	67·5 (10·6)	0·422
Sex			
No. (%) male	53 (64·6)	17 (68·0)	0·757
No. (%) female	29 (35·4)	8 (32·0)	
Site of arrest call			
No. (%) in hospital	36 (43·9)	18 (72·0)	0·014
No. (%) out-of-hospital	46 (56·1)	7 (28·0)	
Delay to CPR (mean (±SD) min)	4·3 (7·0)	0·8 (5·9)	0·019
Delay to intensive care (mean (±SD) min)	11·9 (11·7)	4·0 (6·5)	<0·01

CPR = cardiopulmonary resuscitation. Continuous data are compared using the t-test for independent samples. Frequency data are compared using the Chi-square test.

Table 3 Features of the impedance cardiogram and ECG in Group 1 and Group 2

	Group 1	Group 2	Significance P
Peak dz/dt (mean (±SD)Ωs^{-1})	0·135 (0·074)	0·432 (0·182)	<0·001
Peak-trough (mean (±SD) Ωs^{-1})	0·191 (0·124)	0·678 (0·303)	<0·001
Area 1 (mean (±SD) mΩ)	6·2 (5·0)	25·9 (11·8)	<0·001
Area 2 (mean (±SD) mΩ	5·2 (4·4)	22·4 (10·2)	<0·001
Heart rate (mean (±SD) min^{-1})	85·0 (78·0)	129·1 (39·1)	<0·001

Groups 1 and 2 are compared using the t test for independent samples.

the two ECG/defibrillator pads method (r=0·61, 95% confidence interval 0·43 to 0·83, $P<0·01$). However, the two ECG/defibrillator technique resulted in significantly greater values of Peak dz/dt and Z (1·540 ± SD 0·649 vs 0·908 ± SD 0·192 Ωs^{-1}, $P<0·001$ and 65·7 ± SD 13·9 vs 22·8 ± SD 2·8, $P<0·001$ respectively).

Rhythms

Of the 116 recordings, nine were rejected because of severe motion artefact and/or electrical interference. No recordings were rejected as a result of other adverse physical circumstances. Of the remaining 107 cardiac arrest calls, the rhythm initially recorded was: asystole in 19 (17·8%), ventricular fibrillation in 14 (13·1%), agonal rhythm in 20 (18·7%), electromechanical dissociation in 22 (20·6%), 'shockable ventricular tachycardia' in seven (6·5%), 'non-shockable ventricular tachycardia' in 20 (18·7%) and sinus rhythm in five (4·7%). The mean (±SD) heart rate of the first recorded rhythm and impedance cardiogram-derived parameters for each rhythm are summarized in Table 1 (Fig. 5). For each impedance cardiogram parameter there is a progressive fall in the mean value from sinus rhythm to non-shockable ventricular tachycardia to shockable ventricular tachycardia to each of the pulseless rhythms (i.e. electromechanical dissociation, agonal rhythm, ventricular fibrillation and asystole) (Fig. 5).

Univariate analysis

There was no significant difference between Groups 1 and 2 with regard to age or sex (Table 2). There were significantly more out-of-hospital arrest calls in Group 1 and a significantly greater delay to cardiopulmonary resuscitation and intensive care (Table 2). The mean initial heart rate, where assessable, was significantly greater in Group 2 and each of the impedance cardiogram parameters was greater in Group 2 (i.e. Peak dz/dt, Peak-trough, Area 1 and Area 2) (Table 3).

Multivariate analysis

Using multiple logistic regression the variables age, sex, heart rate, Peak dz/dt, Peak-trough, Area 1 and Area 2 were removed in a stepwise fashion until statistical significance was reached. The two parameters which best predicted a low or absent cardiac output (Group 1) were Area 1 and the Peak-trough measurement (Table 4). Using these two parameters 78 of 82 (95·1%) patients in Group 1 and 20 of 25 (80%) patients in Group 2 were correctly classified.

Discussion

This is the first study to report the effects of cardiac arrest rhythms on the impedance cardiogram in the

Table 4 Multiple logistic regression analysis of the 2 best predictive factors differentiating Group 1 and Group 2 patients

Variable	Coefficient (B)	SE	Wald statistic	Significance *P*
Area 1	0·165	0·067	6·11	0·0134
Peak-trough	4·846	2·556	3·60	0·0579
Heart rate	0·010	0·006	2·83	0·0926
Constant	− 6·379	1·383	21·26	<0·0001

human. To achieve this, it employed a novel electrode configuration in which the impedance cardiogram was recorded through two ECG/defibrillator pads placed in an antero-apical position. It also assessed one established (Peak dz/dt Ωs^{-1}) and three new impedance cardiogram derived parameters (Peak-trough Ωs^{-1}, Area 1 mΩ and Area 2 mΩ) as indices of cardiac function.

The purpose of a haemodynamic sensor for an automatic external defibrillator would be to detect a low or absent cardiac output (the exact measurement is not required). Therefore, the rhythms were divided into two groups. Group 1 consisted of those with an absent or low cardiac output (asystole, ventricular fibrillation, agonal rhythm, electromechanical dissociation and shockable ventricular tachycardia) and Group 2, those with a satisfactory cardiac output (e.g. non-shockable ventricular tachycardia and sinus rhythm i.e. respiratory arrests). On univariate analysis all four impedance cardiogram parameters were significantly lower in Group 1.

The impedance cardiogram has not previously been recorded through two ECG/defibrillator pads. Therefore this method was initially compared with the traditional impedance cardiogram electrode configuration and there was a significant correlation between the values of Peak dz/dt measured using the two techniques. Peak dz/dt was significantly greater when recorded through the ECG/defibrillator pads and this can be explained by the higher baseline impedance associated with the two-electrode technique. Other workers have previously demonstrated that the value of Peak dz/dt is a function of the baseline impedance[22].

In a previous study involving 10 subjects we reported that Peak dz/dt, recorded using the ECG/defibrillator technique, is a highly reproducible value[23].

Use of the Peak measurement of dz/dt as an index of myocardial contractility has been described in the literature. However, in the majority of these references a tetrapolar electrode arrangement employing a longitudinal electrical field was used. It has been significantly correlated with peak aortic blood flow in animals[9,10] and man[11], and has been significantly correlated with stroke volume, stroke work, mean blood pressure, $LVdp/dt_{(max)}$ left ventricular fractional shortening and ejection fraction in animals[10,12–14] and with mean blood pressure and ejection fraction in man[15–17]. Moritz *et al.* is the only study in which a bipolar electrode arrangement was used[18]. In this animal study, they also demonstrated a significant correlation between Peak dz/dt and peak aortic blood flow.

The use of the Peak-trough measurement of dz/dt has not previously been described as an index of cardiac function. However, the peak-trough measurement of ΔZ has been used in the related technique of transcardiac impedance[24,25]. Weiss *et al.* investigated the use of the transcardiac peak-trough measurement of ΔZ to predict the reduction in aortic flow during ventricular fibrillation[24]. In 14 anaesthetized dogs, they measured the transcardiac impedance through two 13·4 cm^2 automatic implantable cardioverter patches attached to the pericardium. Ventricular fibrillation was induced and recordings were made of arterial blood pressure and ΔZ. There was an immediate and very significant fall in ΔZ following ventricular fibrillation induction. In a second experiment, on seven anaesthetized dogs they simulated ventricular tachycardia by rapid ventricular pacing at five different rates[25]. There was a significant correlation between changes in ΔZ and arterial pulse pressure (r=0·89). The authors suggested that ΔZ could be used as a haemodynamic sensor for an automatic implantable cardioverter.

In the development of impedance cardiography, stroke volume was initially calculated from the measured value of ΔZ[26]. Subsequently Kubicek *et al.* demonstrated that an estimate of ΔZ could be obtained from the impedance cardiogram by the product of peak dz/dt and left ventricular ejection time[26]. Left ventricular ejection time can be measured because the B and X points of the impedance cardiogram correspond with aortic valve opening and closing, respectively[19]. However, during arrhythmias it is often difficult to identify these points[27]. Consequently in this study it was not possible to accurately measure left ventricular ejection time. Therefore in order to obtain estimates of ΔZ two integrals of the area under the impedance cardiogram curve were identified which could be easily standardized, even in the presence of arrhythmias. Neither integral has previously been described in the literature. Area 1 was defined as the area under the C wave of the impedance cardiogram curve and above the baseline value dz/dt=0 and Area 2 is the area under the impedance cardiogram curve 50 ms on either side of the peak and above dz/dt=0. On univariate analysis both of these integrals were significantly greater in Group 2 rhythms.

Factors which are significant on univariate analysis may not remain significant on multivariate analysis. Multivariate analysis was therefore performed on the data including age, sex, heart rate and the four impedance cardiogram parameters. The resulting variables used to predict cardiac output were, in order of importance, Area 1 and Peak-trough. Using these two parameters alone the impedance cardiogram correctly classified 78 of 82 (95·1%) Group 1 patients and 20 of 25 (80%) Group 2 patients.

In conclusion, it has been demonstrated that the impedance cardiogram can be rapidly recorded at a cardiac arrest through two ECG/defibrillator pads. It has also been shown that each of four impedance cardiogram parameters i.e. Peak dz/dt (the peak of the impedance cardiogram measured from the line dz/dt=0 Ωs^{-1}), Peak-trough (the peak-to-trough measurement of the impedance cardiogram Ωs^{-1}), Area 1 (the area under the C wave of the impedance cardiogram above the line dz/dt=0 mΩ), Area 2 (the area under the impedance cardiogram 50 ms on either side of the Peak and above the line dz/dt=0 mΩ) were significantly lower in patients with a low or absent cardiac output. On multivariate analysis, the parameters Area 1 and peak-trough best differentiated those with an absent or low output from those with a satisfactory cardiac output. Thus the impedance cardiogram is a potential haemodynamic sensor which should improve the specificity of automatic external defibrillators. Further research is required to prospectively evaluate the impedance cardiogram values derived from this study in a cardiac arrest population and the combination of ECG and impedance cardiogram parameters to improve specificity.

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