33rd Annual International Conference of the IEEE EMBS
Boston, Massachusetts USA, August 30 - September 3, 2011

Detection of Cardiac Arrest using a Simplified Frequency Analysis of the Impedance Cardiogram recorded from Defibrillator Pads

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Abstract— **An algorithm based only on the impedance cardiogram (ICG) recorded through two defibrillation pads, using the strongest frequency component and amplitude, incorporated into a defibrillator could determine circulatory arrest and reduce delays in starting cardiopulmonary resuscitation (CPR). Frequency analysis of the ICG signal is carried out by integer filters on a sample by sample basis. They are simpler, lighter and more versatile when compared to the FFT. This alternative approach, although less accurate, is preferred due to the limited processing capacity of devices that could compromise real time usability of the FFT. These two techniques were compared across a data set comprising 13 cases of cardiac arrest and 6 normal controls. The best filters were refined on this training set and an algorithm for the detection of cardiac arrest was trained on a wider data set. The algorithm was finally tested on a validation set. The ICG was recorded in 132 cardiac arrest patients (53 training, 79 validation) and 97 controls (47 training, 50 validation): the diagnostic algorithm indicated cardiac arrest with a sensitivity of 81.1% (77.6 – 84.3) and specificity of 97.1% (96.7 – 97.4) for the validation set (95% confidence intervals). Automated defibrillators with integrated ICG analysis have the potential to improve emergency care by lay persons enabling more rapid and appropriate initiation of CPR and when combined with ECG analysis they could improve on the detection of cardiac arrest.**

I. Introduction

EMERGENCY pulse checks when cardiac arrest is suspected are inaccurate and can delay cardiopulmonary resuscitation (CPR) by a median of 24s or even longer when no carotid pulse is found [1]. Studies suggest the elimination of the pulse check by lay rescuers [2],[3]. The current International Liaison Committee on Resuscitation / Emergency Cardiovascular Care (ILCOR / ECC) guidelines encourage the initiation of CPR without pulse checks, even when carried out by trained rescuers [4],[5].

Training programs and public access defibrillation (PAD) schemes aim to shorten time from the onset of cardiac arrest until initiation of CPR and shock delivery by ensuring that AEDs (Automatic External Defibrillators) and trained lay rescuers are available in public areas [6]. Automated defibrillator systems that recognize circulatory arrest, using a hemodynamic sensor, together with current algorithms based on electrocardiograms (ECG), would aid in the management of collapsed patients, where accurate, fast and critical decisions must be made. For example, ECG algorithms alone have difficulties to distinguish pulseless electrical activity (PEA) from sinus rhythm (SR). Also, high rate ventricular tachycardia is commonly shocked according to the ECG but it should not be the case if there is no hemodynamic disturbance.

Improved impedance cardiogram (ICG) recording methods have recently enabled their widespread use in critical care [7],[8]. Cromie *et al.* reported both the use of ICG recordings from two defibrillator pads, which overcome the cumbersome application of multiple electrodes, and frequency analysis of the calculated derivative of the impedance signal (dZ/dt) which provided additional information about circulatory arrest in the porcine model [9]. Later they reported a clinical algorithm for detecting cardiac arrest based on the ICG alone using the peak magnitude in a frequency range [10]. They recognized that frequency analysis by Fast Fourier Transform (FFT) in PAD AEDs would compromise its processor (CPU) capabilities and proposed the use of an array of integer filters to estimate the frequency components. In this paper we present the detailed development of these integer filters using the data reported in [10].

II. Materials and methods

A. Equipment

An in-house fully functional defibrillator was constructed (Samaritan AED; HeartSine Technologies, UK) which apart from the ECG, included the recording of ICG using a low amplitude sinusoidal current (30 kHz; 0.05 mA) between the 2 adhesive defibrillator pads (Samaritan, SDE 201, Heartsine Technologies, UK). Its CPU is a Motorola 68336. The ECG and ICG signals were monitored, digitized and stored for retrospective analysis [9],[10].

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B. Clinical Studies

Cromie *et al.* [10] estimated stroke volume using dZ/dt_{max} which is the maximum negative deflection of the first order derivative of the impedance Z. Records were initially analyzed in 5s windows and dZ/dt_{max} was determined at a bandwidth of 1-11Hz. The dZ/dt was also analysed by the FFT and the peak magnitude (PM), from the frequency components between 1.5 – 4.5Hz, was identified as a marker for cardiac arrest. Additionally multivariate analysis was carried out to fit stroke volume with linear and quadratic terms for both dZ/dt_{max} and PM.

The ICG was recorded in 132 Cardiac Arrest patients attending within hospital and in the community, by an Emergency Medical Team (EMT) from 1st May 2003 – 31st December 2004. The modified Samaritan AED was used with adhesive ECG+ICG defibrillator pads applied in standard cardiac arrest positions (inferior to right clavicle in midclavicular line, to right of upper sternum and over left lower chest). The study complied with the Declaration of Helsinki. Ethical approval was obtained from the Local Regional Ethical Committee. All survivors and the next of kin of non-survivors were informed retrospectively by a letter, and could withdraw consent for the use of their data. No interruption was made in patient management during this study.

In addition, 97 Controls provided the Non-Arrest records for a sustained circulation. This group included patients with a low cardiac output and cardiac symptoms referred for myocardial perfusion imaging.

The output status from the ECG, using EMT documentation, was identified by an investigator blinded to the ICG. SR, atrial fibrillation (AF), ventricular tachycardia (VT; pulseless and non-pulseless), ventricular fibrillation (VF), PEA, agonal rhythms and asystole were noted. Ambiguous rhythms with artifacts or incomplete documentation and no identified cardiac output status, were labeled 'Miscellaneous'. ECG and ICG records distorted by artifacts caused by CPR were excluded from the analysis.

Consecutive data were divided into Training and Validation sets. The Training set consisted of 53 Cardiac Arrest patients and 47 Non-Arrest patients. The balance between sensitivity and specificity detecting cardiac arrest was used to determine the best cut off for the algorithm on the data from the Training set. A specificity higher than 99% and sensitivity greater than 84% (a reference value for untrained first responders [1]) was considered acceptable. The algorithm for detection of cardiac arrest was tested with unseen data provided by the Validation set which contained 79 Arrest patients and 50 Non-Arrest.

C. Filters

Real-time FFT analysis could compromise the processor power of a defibrillator. Therefore an alternative approach using an array of integer filters was adopted for frequency analysis. The aim is to determine if there is a dominant frequency in the signal that could indicate an organized rhythm suggesting that there is no cardiac arrest. This alternative approach to FFT is not aimed at achieving accuracy in the calculation of the frequency spectra but to answer the question if a dominant frequency can be identified using less processing power or at least spreading the load of the calculations during the use of the defibrillator. Each dZ/dt trace, both for Cardiac Arrest and Control patients was initially analyzed in 6-second epochs (1024 samples). The dZ/dt signal was passed through multiple band-pass integer filters at frequency steps of 0.1666Hz ($170.667/1024$) over a range of 1—5 Hz. From the training set a subset of 19 cases: 13 cardiac arrest and 6 normal sinus rhythm (controls) were used for the design of the integer filters to replace the FFT process.

Different implementations of integer filters were tested. In general for integer filters, the implementations are designed to particular frequencies, for instance fractions of Nyquist frequency [11] or they do not allow fine detail for the required bandwich [12]. The filters were initially tested using sinusoidal signals generated at different frequencies. Simplicity and a relative low leakage were the initial criteria for filter design optimization. Posterior comparisons with FFT methods on the ICG were considered. The Root Mean Square (RMS) of the filtered signal was taken as the amplitude of the frequency component in observation. The correlation of the RMS amplitudes from the filter and magnitudes from the FFT, together with the ability for identifying the peak magnitude for cardiac arrest detection within the interval 1—5 Hz, when compared to FFT, were the final criteria for the filter design.

Simple bandpass filters were employed [13] and integer approximations were used. Two poles are placed as close as possible to the unit circle at an angle given by $\theta = 2\pi(fc/fs)$ and two zeros at z = ± 1. fc and fs are the cutoff frequency and the sampling frequency respectively. In figure 1, the unit circle and the position of the two poles and the two zeros for the passband filters is depicted.

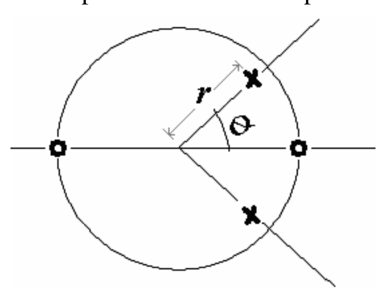

Fig. 1. Positions of the poles (**x**) and zeros (**o**) for a passband filter, $0 < r < 1$.

The corresponding transfer function is as follows:

$$H(z) = \frac{1 - z^{-2}}{1 - (2r\cos\theta)z^{-1} + r^2 z^{-2}} \qquad (1)$$

For which the difference equation is given by:

$$y[i] = (2r\cos\theta)y[i-1] - r^2 y[i-2] + x[i] - x[i-2] \qquad (2)$$

Initial experiments were implemented to set the minimal scales that provided accurate results. For integers only, scaled cosines were values between 0 and 10000, and the radius r set to 0.99 was scaled to 99. Gain was 100. However, for further speed efficacy in relatively low specification CPUs, bit shifts instead of divisions were implemented. Thus cosine values in (2) were approximated by rational numbers of the form $c/2^{13}$, where $c \in \{0,\ldots,2^{13}\}$ and r was set to $127/2^7$. The estimation of the frequency spectra can be calculated on a sample by sample basis. For each analyzed frequency and for an arriving sample 4 multiplications, 6 adds and 6 bit shifts are required for the implementation of (2) and the calculation of RMS. Implementations of filters and algorithms are in the C language. They were initially designed and tested in a PC and posteriorly embedded into the device.

D. Algorithm for detecting Cardiac Arrest

The algorithm based on the ICG to assess cardiac arrest for each epoch (figure 2) uses both the normalized peak magnitude converted to decibels and squared (PM^2) and the RMS of the unfiltered dZ/dt signal (R_0). The conversion to decibels could be omitted (updating accordingly the thresholds and evaluations in the flow diagram). However it is presented since PM^2 was drawn from the multivariate analysis described in [10]. A low PM^2, in keeping with the presence of a strong frequency component, indicates that there is no cardiac arrest.

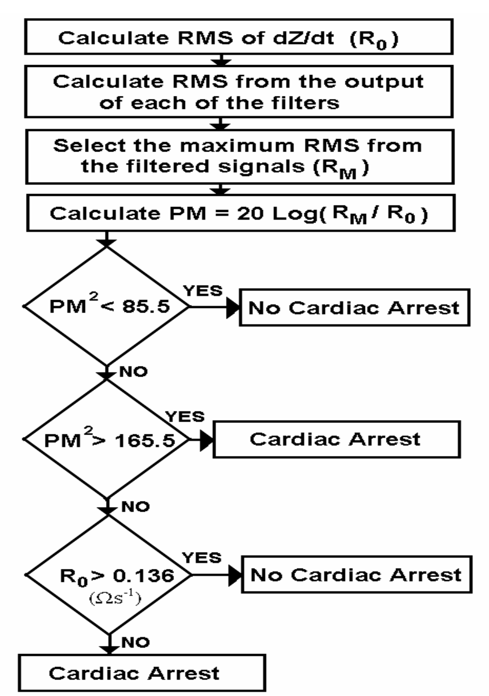

Fig. 2. Flow diagram to detect cardiac arrest using peak magnitude (PM) and RMS of dZ/dt for a 4s epoch.

Additionally, epochs were analyzed in conjunction with its two preceding epochs as 3 consecutive 4s epochs. The diagnosis of circulatory arrest was finally confirmed or refuted by the algorithm when similar results were obtained in 2 of the 3 consecutive epochs. If only 2 consecutive epochs were available they were excluded from the triplicate analysis. The output from the diagnostic algorithm during each consecutive triplet of 4s epochs (12s) was compared with the original clinical diagnosis in the presence or absence of cardiac arrest, as determined from the ECG and medical documentation.

III. RESULTS

Frequency response for one of the proposed bandpass filters can be seen in figure 3. For a cutoff frequency of 3Hz the maximum is slightly shifted to the right and the magnitude is higher than –3db for frequencies between 2.85Hz and 3.23Hz. For a frequency f_T between 1 Hz to 5 Hz their magnitudes are higher than —3db for [f_T -0.15, f_T +0.23]Hz approximately.

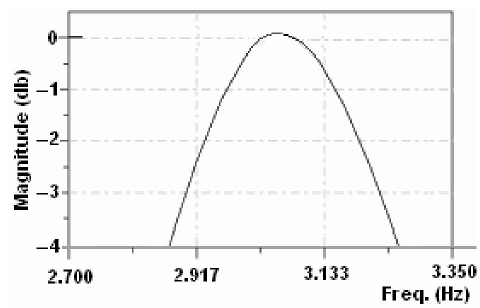

Fig. 3. Frequency response for the proposed bandpass filter of 3Hz.

An estimation of the spectra of frequencies obtained by the filters was compared to FFT results from 19 cases of ICG signals. Table I shows average peak magnitude from FFT and average peak amplitude from filters calculated for frequencies from 0 to 85.333Hz (Nyquist $170.667/2$) at a step of 0.1666Hz. Blocks of 6s (1024 samples) were used.

TABLE I
COMPARISON OF PEAK MAGNITUDE (PM) BY FFT AND FILTERS FOR CARDIAC ARREST AND SINUS RHYTHM (SR)

Rhythms		PM (FFT) db	PM (Filters) db
Cardiac Arrest	average	-33.0830	-19.4755
(13 cases)	std	2.4522	3.4509
SR	average	-22.4686	-7.3571
(6 cases)	std	4.4568	4.3586

TABLE II
RESULTS FOR THE DETECTION OF CARDIAC ARREST IN TRI-EPOCHS

	Correctly Identified	Incorrectly Identified	Sensitivity CI 95%	Specificity CI 95%	Total
Training					
Arrest	320	39	89.1% 85.4-92.1		359
Non-Arrest	4789	20		99.6% 99.4–99.8	4809
Validation					
Arrest	455	106	81.1% 77.6–84.3		561
Non-Arrest	8828	267		97.1% 96.7–97.4	9095

Correlation coefficients between magnitudes from the FFT and RMS from the filtered signal for each frequency are in general higher than 0.6 for the studied cases. Results for the algorithm to detect cardiac arrest using 3 consecutive epochs [10] are shown in table II.

IV. Discussion

The key for the algorithm presented is the possibility of estimating the magnitude of the dominant frequency for dZ/dt in relatively limited capacity CPUs. Integer FFT can be implemented since it uses only adds and bit shifts however it presents inconveniencies. It requires gathering the total number of samples to start the calculations, demanding too much CPU resources *from* that particular time. Instead, the proposed bank of filters spread the load for the CPU as the estimation is carried out sample by sample. This feature is convenient for the defibrillator since the CPU is constantly dealing in real time with input/outputs, running diagnosing algorithms and the possibility of delivering shocks. The sample by sample approach also facilitates the use of different window lengths in order to optimize this parameter for the diagnosis by the algorithm. Initially it was set to 1024 samples to cover 6s which facilitated the comparison to the FFT and the design of the filters. Later this window length was finally set to 4s (683 samples) which provided an optimal window of analysis in a fully functional defibrillator.

The number of filters at particular frequencies which are interrogated can be fixed (and reduced) depending on the application in which the bank of filters is implemented. This feature provides savings in CPU use since the number of operations is proportional to n, the sample size, and for large n it represents an advantage over the $O(n\log n)$ FFT. However, for the application described in this paper this advantage was not fully exploited due to the relatively low n.

An algorithm to detect cardiac arrest based on the estimation of the magnitude of the dominant frequency of dZ/dt has shown promising results. The strategy of analyzing 3 consecutive 4s epochs provided better results than the analysis of one epoch. For the validation set, when compared to untrained first responders [1], the sensitivity is marginally lower than 84% however the specificity of around 97% is much higher than 36%. The approach presented does not use the information provided by the ECG signal. The aim for this development was to study the possibilities that the ICG alone can offer to determine different kinds of cardiac arrest. It is less efficient for diagnosing VF than current ECG based algorithms [10] but offers opportunities for improvement in other situations. For example, an ICG based algorithm coupled to an ECG algorithm could be more efficient for detecting PEA from SR and to consequently advise about CPR [8]. Another application would be to distinguish pulseless from non-pulseless VT.

V. Conclusion

Estimation of the frequency spectrum of the first order derivative of the impedance cardiogram (dZ/dt) recorded through the 2 transthoracic defibrillator pads can be used as a marker of circulatory collapse. The use of less accurate integer filters for the estimation is a feasible solution to be applied in a less powerful CPU operating in a PAD defibrillator. The results provide initial tools for further development of applications for the use of ICG in defibrillators during emergency clinical practice.

Acknowledgment

The authors thank Audrey Tomlinson RGN and Leslie Swales RGN of the Royal Victoria Hospital, Belfast, for their help in data acquisition and Heartsine Technologies, for help and support with equipment and data analysis.

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