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Cardiac Frequency Study During Ventricular Fibrillation Using R Peak Location Slope

Abstract. *The present paper introduces a new method for cardiac frequency estimation directly from the positions of R peaks, this work aims to present and interpret a novel method based on the slope of the curve that reproduces the positions of the R peaks versus their respective indices, which is used to assess differences in RR time series dynamics in patients during ventricular fibrillation. The goal of this technique is to evaluate the cardiac frequency during normal and ventricular fibrillation beats through a visual inspection of the change in the heart rate. The main purpose is to verify the relationship between the slope and the change in the cardiac beat type. The greatest advantage of the proposed method is to recognize the time of the onset of the ventricular fibrillation by simply referring to the change in the slope. Therefore, it is necessary to start with a QRS complexes detection algorithm to find the position of R peaks. The evaluation of this technique is performed using the Creighton University ventricular tachyarrhythmia standard database (CUDB).*

Streszczenie. *W niniejszej pracy przedstawiono nową metodę szacowania częstości akcji serca bezpośrednio z pozycji pików R. Celem tej pracy jest przedstawienie i interpretacja nowatorskiej metody opartej na nachyleniu krzywej odtwarzającej położenie pików R w funkcji ich odpowiednich wskaźników, co służy do oceny różnic w dynamice szeregów czasowych RR u pacjentów z migotaniem komór. Celem tej techniki jest ocena częstości akcji serca podczas uderzeń normalnych i migotania komór poprzez wizualną kontrolę zmian częstości akcji serca. Głównym celem jest sprawdzenie związku pomiędzy nachyleniem a zmianą typu rytmu serca. Największą zaletą proponowanej metody jest rozpoznanie czasu wystąpienia migotania komór poprzez proste odniesienie się do zmiany nachylenia. Dlatego konieczne jest rozpoczęcie od algorytmu wykrywania zespołów QRS, aby znaleźć położenie pików R. Ocenę tej techniki przeprowadza się z wykorzystaniem standardowej bazy danych tachyarytmii komorowej Uniwersytetu Creighton (*CUDB). **((*Badanie częstotliwości serca podczas migotania komór przy użyciu nachylenia położenia szczytu R*)**

Keywords: Electrocardiogram, R peak detection, ventricular fibrillation, slope, cardiac frequency, heart rate.
Słowa kluczowe: Elektrokardiogram, wykrywanie szczytu R, migotanie komór, nachylenie, częstość akcji serca, częstość akcji serca.

Introduction

The cardiovascular disease is the most common cause of the half deaths in the world during the last decade. Accordingly, the diagnosis and treatment of these hazardous conditions seem to be a vital task. In cardiology, the Electrocardiogram (ECG) signal is still one of the most prevalent and widely used tools for diagnosing and analyzing cardiac arrhythmias. The ECG exam is in fact a non-invasive tool done by the doctor to explore the functioning of the heart by using external electrodes that come into contact with the skin. This signal is a reflection of the electrical activity of the heart and brings together three major waves: P, QRS and T, in addition to certain intervals and segments. Generally, durations and shapes of different wavelengths are considered telltale signs of certain cardiac anomalies [1, 2].

One of the main causes of sudden death in patients with heart disease is Ventricular Fibrillation (VF). It is a malignant arrhythmia characterized by a fast heartbeat and uncoordinated contraction of the cardiac muscle of the ventricles in the heart [3, 4, 5, 6]. The VF is usually diagnosed from the patient's ECG data. It takes the form of a sinusoidal signal with an irregular shape and pulses of varying amplitude (Fig. 1). The heart rate in such a situation can be between 240 and 600 beats per minute (bpm) or more [7]. The cardiac frequency increases or slows down depending on the effort, emotion, etc. At rest, this may drop to 45 bpm, while in a state of fever or emotion, it can exceed 100 bpm. During exercise, the heart rate is directly related to the intensity thereof, a maximum effort will accelerate the heart rate to 180 bpm. So, the distinction between normal variation and arrhythmia cannot be strict, except for very high frequencies.

The aim of this work is closely related to the calculation of the cardiac frequency from the detection of the QRS complex and the Heart Rate Variability (HRV). The location of these QRS complex is obtained by the use of a detector algorithm based on the method suggested by Pan and Tompkins [8]. The calculation of the heart rate allows us to discriminate the rhythm change or the frequency; it is carried out by calculating an instantaneous frequency based on the slope of the curve that produce the positions of R peaks versus their respective indexes. Thus, we can extract an average frequency given by the slope of the line. The existence of multiple slopes in the same plot indicates the rate of change in the given signal. For validation, these techniques are applied at Creighton University ventricular tachyarrhythmia database (CUDB); it includes many episodes of ventricular fibrillation [9, 10].

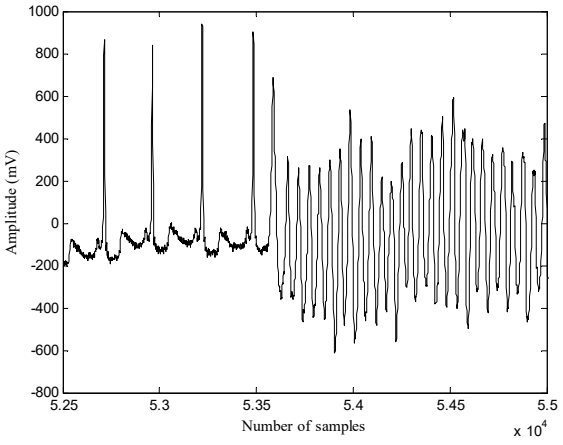


Fig. 1. Onset of VF in a recorded ECG

Methodology

Clinical data

The database selected to test the efficiency of the method was from the Creighton University ventricular tachyarrhythmia database (CUDB), because these records

contain episodes of VF in nearly all its records. This database was collected by Dr. Floyd M. Nolle at the Creighton University Cardiac Center as part of his work on ventricular fibrillation in the surface electrocardiogram. It includes 35 single-channel records, each of which shows the onset of ventricular fibrillation. All signals were passed through an active second-order Bessel low-pass filter with frequency cutoff of 70 Hz, and were digitized at 250 Hz with 12-bit resolution over a 10 V range (10 mV nominal relative to the unamplified signals). Each record consists of 127232 samples (somewhat less than 8.5 minutes). In episodes of heart failure, the fibrillation is almost always preceded by a series of ventricular tachycardia, which eventually yields to the fibrillation itself. The reference annotation files provided in this database have been included to assist users in locating events of interest. All beats are labeled normal, N, (although many are ectopic). The VF onset annotations mark only the approximate beginnings of the VF episodes. So, in order to be as rigorous as possible our complete test database consisted of 4 records only [9, 10].

QRS complex detection algorithm

The algorithm used here is based on a modification of the QRS detector proposed by Pan and Tompkins (PT). It locates the QRS complexes in ECG signals using bandwidth filter, slope and pulse time criteria [8]. Pan and Tompkins (PT), also known as the low-pass differentiation algorithm (LPD), introduced, in 1985, a major advancement in ECG signal processing. The QRS detection was implemented through three detection stages: linear digital filtering, non-linear transformations, and decision rule. The algorithm applies a special digital band-pass filter, which can reduce false detections due to different types of interference contained in the ECG signal. In summary, the QRS complex detection algorithm consists of the following processing steps as shown in Fig. 2:

- band-pass filtering (a cascade of low-pass and high-pass filters),
- differentiation,
- squaring,
- moving window integration,
- decision rule: adaptive thresholds are used.

Because the QRS complex amplitude can vary significantly within a short interval of time, requiring an adaptive detection, the threshold is calculated using a 10-point iterative estimation; the algorithm automatically adjusts the thresholds and the parameters to accommodate the changes in the QRS morphology; the peaks that pass the threshold are marked as an R wave. Moreover, in most cases, if no QRS complex was detected within 150% of the RR interval duration estimated by a 10-point iterative estimator, then by dividing the threshold by two, the search is performed back [11].

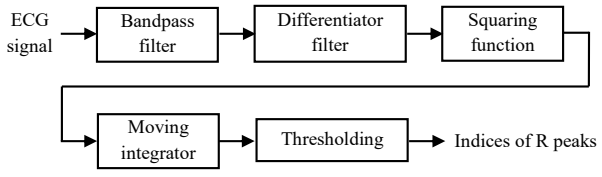


Fig. 2. The functional scheme of the Pan and Tompkins algorithm

Heart rate – Cardiac frequency

The heart rate or the cardiac frequency (CF) can be measured by the pulse duration. The heart rate is defined as the number of cardiac pulsation (beats) per minute. The signal of the heart rate variability (HRV) represents the temporal evolution of the heart rate; it is obtained starting from the extraction of the time intervals between the R peaks of the ECG signal. It corresponds to the time (in milliseconds) which separates two ventricular contractions one from the other.

Therefore, more this time is high more the CF is low, and inversely. The analysis of the HRV can be carried out in the temporal field by graphically representing the RR intervals according to time or according to the irrespective indices, which gives a rise to curve called tachogram (Fig. 3). The analysis of the tachogram is one of the bases of the study of the HRV [12]. A strongly variable tachogram indicates a very irregular rhythm; on the other hand, small variations are sign of stability of heart rate.

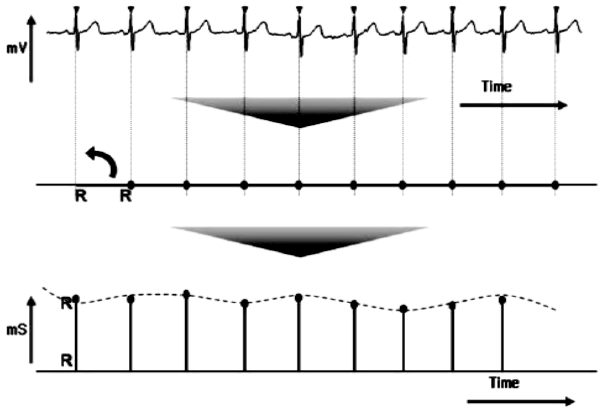


Fig. 3. RR intervals and their fluctuations

Computation of the cardiac frequency

The computation of the cardiac frequency consists, initially, in calculating the time interval between two successive identical waves. In an ECG waves we are usually dealing with intervals such as P-P, Q-Q, R-R, S-S or T-T. By preoccupation with a facility of detection, we choose, most of the time, to calculate the RR interval to determine the cardiac frequency. For this reason, the method consists in isolating the QRS complex from the ECG, detecting the R peak then calculating the time which separates it from the last R peak detected. We can then determine the value of the RR interval as being the time passed between the last two detected peaks:

$$RR = R_i - RR_{i-1} \tag{1}$$

Then, we calculate the cardiac frequency according to this relation:

$$F_C = \frac{60}{RR} \tag{2}$$

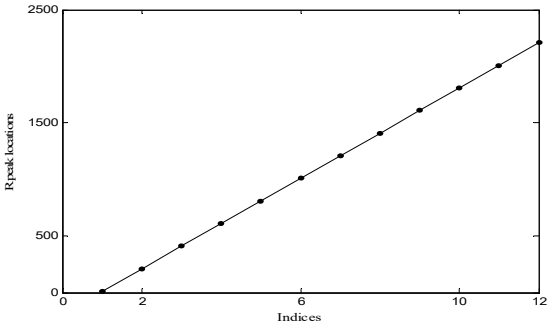
Where RR is given in seconds and F_C in beat per minute (bpm).

Technique based on the slope

The technique is based on the positions of the R peaks detected previously. Normally, the position of the R peaks forms an increasing set of number: from the first one to the last one. The layout of these positions according to their respective indices generates a curve which increases according to the index. This is another manner of observing the variability of interval RR, since each elementary slope indicates the difference between two consecutive positions. Therefore, if the distance which separates two successive positions (RR interval) is quasi constant, we are dealing with a very regular rhythm which appears by a linear layout of the positions (Fig. 4).

In this case, the regularity of the rhythm is characterized by the slope of the line which will be constant. However, if this

distance varies obviously, the curve will present a break line which will present many slopes according to the variability of the rhythm (Fig. 5). Therefore, the analysis of the behavior of the slope informs us about the changes and the variability of the heart rate. A slope which tends towards the infinite indicates the beginning of a bradycardia



(deceleration of the rhythm). On the other hand, if the slope converges towards 0, it means that a tachycardia starts (acceleration of the rhythm).

Fig. 4. Quasi-constant distance which separates two successive R peak locations

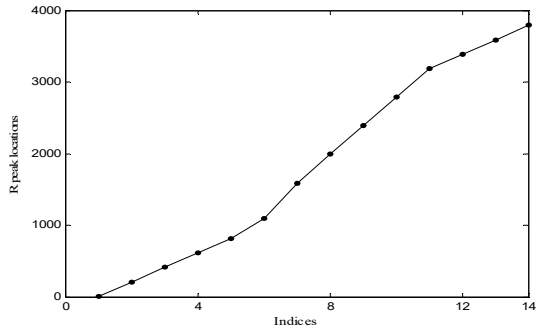


Fig. 5. Irregular distance which separates two successive R peak locations

Application, results and discussions

R wave detection stage

For the evaluation of the detection algorithm, the technique is performed using the CUDB database. According to the work in [13], the detected R-waves were compared to the annotation file attached to each signal to determine the Error Rate (ER) given by:

$$ER(\%) = 100 \times \frac{FP + FN}{TNB} \tag{3}$$

Where: TNB is the Total Number of Beat in the annotation file, FP is the number of False Positive or false position detected by the algorithm and FN is the number of False Negative or missed beat. Then, the Detection Rate (DR) is given by:

$$DR(\%) = 100 - ER(\%) \tag{4}$$

Nevertheless, the VF episodes were excluded from the detection process because we have only considered ECG rhythms that are annotated in the database. We have selected a total of 4 records; all channels were included having a total of 2201 beats. The numerical values of the R wave detection are given in table 1. It show high efficiency and the results of the R wave detection were greatly promising for the cardiac frequency evaluation. However, we can also see that for some records only modest detection results were obtained by the algorithm. This is, in general, mostly due to the very high noise level found in these signals causing in many FN.

Table 1. R-wave detector test results applied on the CUDB and using the PT algorithm

Record	TNB	FN	FP	ER (%)	DR (%)
cu01	203	3	0	1.48	98.52
cu03	930	5	9	1.51	98.49
cu05	693	14	7	3.03	96.97
cu07	375	0	0	0	100
Total	2201	22	16	1.73	98.27

Method of the slope

In this section, we applied the PT detection way on the entire record including the VF episodes. Once the positions of the R peak are available, this series is traced in a reference mark where the x-axis represents the index of the ordinates that contains the positions. For the signals chosen, Fig. 6 illustrates the layouts of the R positions. The slopes are calculated by applying the linear approximation “cftool” (curve fitting toolbox) of MATLAB [14]. By visual inspection, we can mention the clear change of the slope for all the curves. By taking the example of the Fig. 6-a, the signal cu01 presents a transition which is at the point of index 205, this corresponds to the position 53930.

Therefore, if one returns to the annotation file (cu01.atr) the change of rhythm, which is the beginning of a VF, starts approximately at the sample 53541, which constitutes a point of discrimination between Normal (N) and VF rhythm. For the other signals, the results show lines of different slopes mentioning the transitions between a normal rhythm and a VF rhythm. The modified points from the slope are mentioned on the figure, they are compared with those given by the file annotations of the rhythm.

Tables 2 and 3 summarize the values of the various slopes as well as the indices of transition between the rhythms. These indices are drawn directly from the figures as first step and drawn from the files of annotations as second step. Concerning the value of the frequency, we have noticed an increase in this one during the phase of ventricular fibrillation. In the case of cu07 record, we recorded a value of the cardiac frequency that exceeds 800 bpm. This is due, possibly, to the R peak detection algorithm causing a high number of waves identified in the ventricular fibrillation area which is characterized by its variable data. However, most of the recorded scores lead to frequencies around 300 bpm for VF episodes.

Conclusion

In this work we have developed an innovative method for evaluating the heart rate or the cardiac frequency in the case of ventricular fibrillation, the method depicted forward in the present work exploits the regularity observed in the R peak positions. This technique is applied at Creighton university ventricular tachyarrhythmia (CUDB) Database; it is based on the locations of the detected R waves, or any waves corresponding to the VF episodes, by plotting the minimum reference mark giving the position according to its index, and the different average frequencies, which are expressed by the slope of the linear fit. Therefore, we have reached the raised fibrillation which is naturally characterized by frequencies equal to 450 bpm and it corresponds to a slope that tends towards zero. But, if the selected recordings have results extremely convincing, probably it is not the case for all the files of the CUDB due to the very large variations in these recordings. Finally, the slope method that gives an efficient visual discrimination algorithm for the normal and VF cardiac rhythms can be used to compute the cardiac frequency in very practical manner.

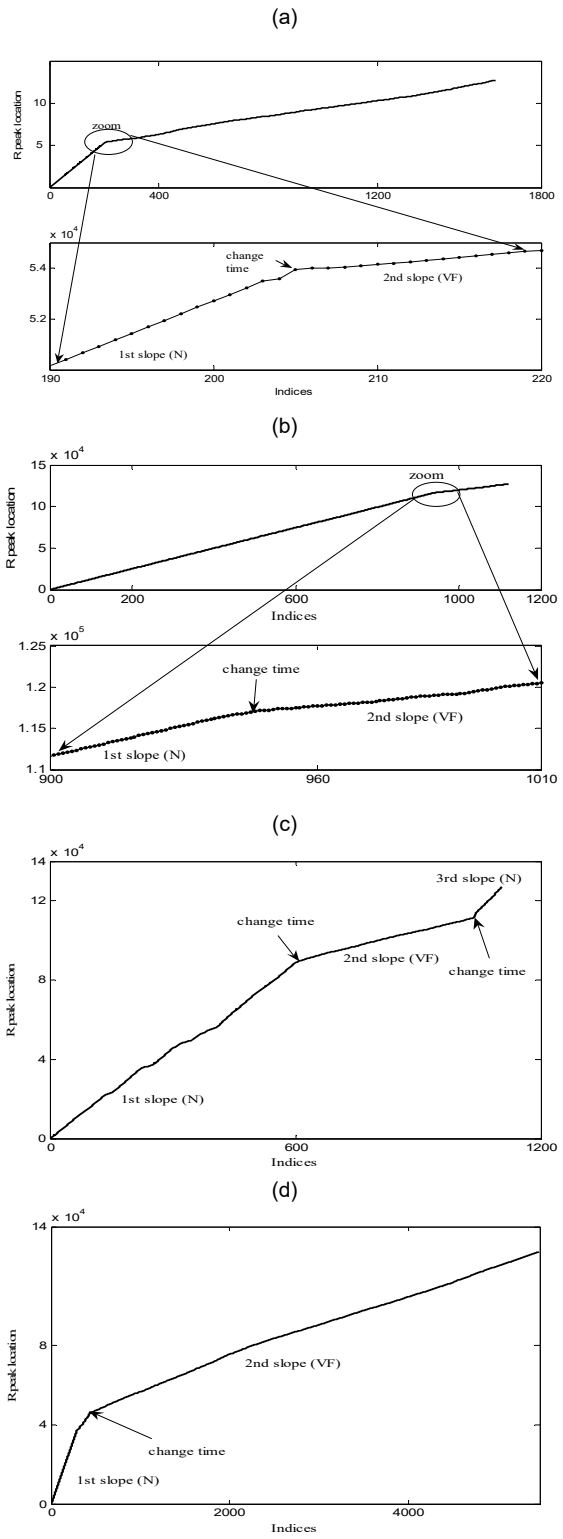


Fig. 6. The slope method applied to the cudb: (a) cu01.dat, (b) cu03.dat, (c) cu05.dat, (d) u07.dat

Table 2. Various slopes (in second) and frequencies (in bpm)

Record	Slopes (sec.)			freq. (bpm) = 60/slope		
	1	2	3	f_1	f_2	f_3
cu01	1.06	0.197	–	56.6 (N)	304.6 (VF)	–
cu03	0.49	0.193	–	122.4 (N)	310.9 (VF)	–
cu05	0.57	0.2	0.846	105.3 (N)	300 (VF)	70.9 (N)
cu07	0.443	0.068	–	133.4 (N)	882.4 (VF)	–

Table 3. Indices of transition between the rhythms (in second)

Record	Change time (in figure)	Change time (in atr file)
cu01	53930 (215.7 s)	53541 (214.2 s)
cu03	116930 (467.7 s)	116430 (465.7 s)
cu05	89651 (358.6 s) & 111583 (446.3 s)	89692 (358.8 s) & 111598 (446.4 s)
cu07	45512 (182.05 s)	45502 (182.01 s)

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