A QRS Complex Detection Algorithm using Electrocardiogram Leads

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Abstract

A QRS complex detection algorithm was developed using the available leads of the electrocardiogram (ECG). This detector is based on the combination of two improved versions of QRS detectors available in the literature. An important characteristic of this algorithm is the possibility of using two or more ECG channels for QRS detection. The first detection method is based on a cross number in a detection threshold defined by the authors. When a low reliability situation occurs in the first method, the output of the second detection method is used to confirm or reject the detection. The second method also uses an adaptive detection threshold defined by the authors and a candidate QRS is tested against some criteria that use features as amplitude, width and RR interval to validate the candidate as a QRS. Testing the algorithm with MIT/BIH Arrhythmia Database resulted in 99.22% sensitivity and 99.73% positive predictivity.

1. Introduction

The detection of QRS complexes in real time is an important function of cardiac monitors.

This study developed an algorithm to detect QRS complexes which combines two detection modules based on methods available in the literature. Such combination allows the use of the best features of each detection module and eliminates their specific weaknesses.

A great hurdle to be overcome in a reliable detection is the sensitivity of the electrocardiogram to several disturbance sources such as powering source interference, movement artifacts, baseline wandering and muscle noise. Two or more channels of the electrocardiogram (ECG) are used in order to improve reliability in detecting QRS by reducing the impact on detector performance if one of the channels presents low quality signals (low amplitude or contamination by noise and artifacts).

2. Methodology

The database of the Massachusetts Institute of Technology and Beth Israel Hospital (MIT/BIH) was used for development and analysis of the algorithm. This database has 48 records which present two electrocardiogram leads, each record lasting 30 minutes. 44 records of the database were used in the development of the algorithm, in compliance with the Association for the Advancement of Medical Instrumentation - AAMI/EC57/1998 [1] which suggests the exclusion of the 4 records with paced beats in cases where the QRS detector does not have the specific algorithm to detect this type of beat.

The records were altered to fit the signal characteristics of the ECG monitor in which the detector would be applied. The software *xform*, available in the MIT/BIH database CD-ROM, was used for this purpose. Sampling frequency was altered from 360Hz to 250Hz, signal gain from 200 adu/mV (analog to digital unit per miliVolt) to 160adu/mV, signal resolution from 11 bits to 12 bits and baseline from 1024 adu to 2048 adu.

Figure 1 presents the block diagram of the QRS complex detector, composed by a signal conditioner stage and two QRS detector modules. It is based on two distinct detection methods executed in parallel with complementary characteristics in order to make the QRS detection more reliable.

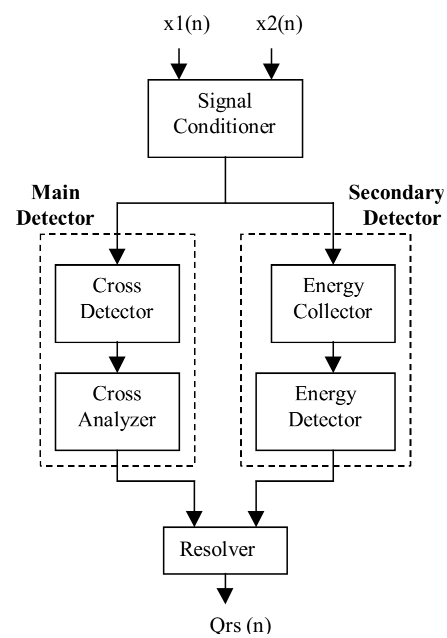

Figure 1 – Block Diagram of the Detector

2.1 Signal Conditioner

The first stage, the signal conditioner, filters and fits the sampled signals from two ECG channels generating the signal which will be used in the latter stages. Figure 2 represents this conditioner, constituted by two channels with three stages of filtering in cascade, a calculator of the absolute value of the output from the third stage of filtering and an adder of each channel's output which will have as output the signal that will be used in latter stages.

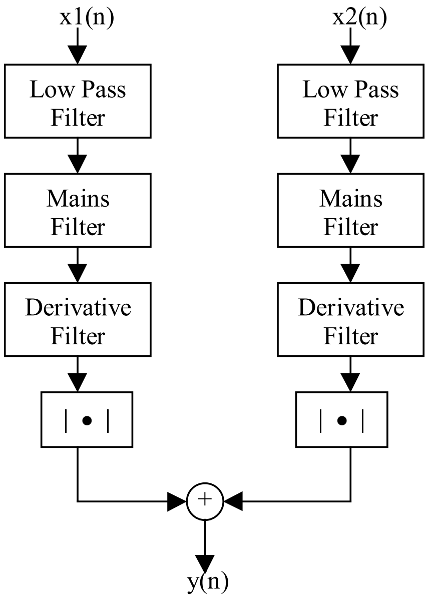

Figure 2 – Block Diagram of the Signal Conditioner

The first filtering stage is constituted by a low band pass filter, described by the equation:

$$4y(n) = x(n-2) + 2x(n-1) + x(n), \quad n \in N \qquad (1)$$

The second filtering stage is constituted by a notch filter; its zero was calculated so it would attenuate the first harmonic of the interference from the powering source, described according to the equation:

$$y(n)=x(n-2) - 2\cos\left(\frac{60\pi}{125}\right)x(n-1) + x(n), \, n \in N \qquad (2)$$

The third stage is constituted by a derivative filter, described by the equation:

$$y(n) = x(n) - x(n-6), \quad n \in N \qquad (3)$$

Figure 3 presents the gain in frequency response of the filter which is a result of the three filters in cascade. This filter presents a band pass between 9 and 30 Hz, with 17 Hz of maximum gain, covering a spectrum range in which an ECG signal not contaminated by noise presents highest energy[2].

The stage after each signal conditioner channel simply calculates the absolute value of the output signal of the corresponding filter. The arithmetic mean of the absolute values for each channel will be used by the latter stage of the QRS detector. The calculation of the absolute value of the signals was used in order to avoid possible canceling out of the ECG signals.

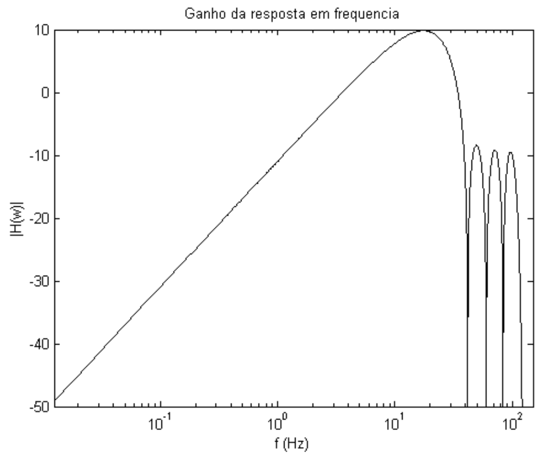

Figure 3 – Gain in frequency response of the result from the filter cascade.

2.2 Main Detector

The main detector is based on the one proposed by Engelse and Zeelenberg [3] and uses as criterion to determine the occurrence of a QRS complex the number of times the input signal rising edges cross a threshold in a determined time interval.

The event analysis is done if there is at least one crossing followed by 180 ms without a new crossing or if there are more than 4 crossings.

If there are more than 4 crossings the event is classified as *noise*.

If there are 2 to 4 crossings, the event is classified as a *QRS Complex.*

If there is only one crossing, there is an undefined event and the output from the secondary detector will be used in order to determine whether the event is a QRS. The threshold used to determine the number of crossings is continuously adapted according to input signal amplitude. Two thresholds are used for this purpose: the baseline threshold (BT) and the detection threshold (DT). The BT is not used for comparisons, but as basis for the DT settings every time an event is detected. Therefore, the BT must vary according to the amplitude of the last signals and verify their tendency. The DT, the one actually used for comparison, has to be more variable, since it has to reach high values only just immediately after an event to avoid erroneous detection of T waves. After this period, however, the threshold can be readjusted to a low value, but not so low, in order to avoid the detection of false positives.

THRESHOLD CALCULATIONS

When there is noise:

$BT \Leftarrow 1,5 \times BT$ (4)

$DT \Leftarrow \max(05 \times (QRS\ peak), BT)$ (5)

When the critical time is reached without any occurrences of a QRS complex:

$BT \Leftarrow 0,5 \times BT$ *(6)*

$DT \Leftarrow BT$ *(7)*

When a QRS is detected:

$BT \Leftarrow (0,75 \times BT + 0,25 \times (QRS\ peak))$ (8)

$DT \Leftarrow \max(0,5 \times (QRS\ peak), BT)$ (9)

There is a second critical time which assumes maximum value after the occurrence of a QRS complex or noise. When this time is reached without any crossings, its value is reduced and the DT is readjusted:

$DT \Leftarrow 0,75 \times DT$ (10)

$DT \Leftarrow \max(DT, 0,5 \times BT)$ (11)

Maximum (1200) and minimum (100) values were determined for BT and DT in order to guarantee minimum sensibility to avoid the detection of noise as a QRS complex and to avoid that artifacts or occasional QRS complexes of high amplitude make the detector insensible to subsequent complexes.

The described method was chosen to implement the algorithm of the main detector because of its high sensitivity. However, when there is only one threshold crossing, there is an undefined event which could be caused by a QRS complex or by baseline wandering or by electrode movement artifact. When this situation occurs the result of the secondary module is used to determine if this event is a QRS complex.

2.3 Secondary Detector

This detection stage is based on the detectors proposed by Pan and Tompkins [4] and Ligtenberg and Kunt [2]. It is constituted by an energy collector stage and by the detector itself (Figure 1). It analyses the energy signal of the ECG since their characteristics allow for a good characterization of QRS complexes, more facility in avoiding noise detection and identifying baseline wandering and movement artifacts.

The energy signal is obtained after the input signal of the main detector module passes through a moving window integrator, corresponding to the energy collector and described by the equation:

$$20\,y(n) = \sum_{i=0}^{19} x^2(n-i), \quad n \in N \qquad (12)$$

This detector module presents a detection threshold based on the article by Ligtenberg and Kunt [3], obtained from an adaptative process where threshold settings are determined empirically.

In order to raise detector predictivity, i.e., avoid the detection of false positives, features such as RR interval, amplitude and width are used to characterize a QRS complex.

The detection threshold setting, determined empirically, is used to make the algorithm very sensitive in order to detect the highest number of QRS complexes possible and at the same time avoid false positive detections.

Therefore, the following rules for threshold settings were obtained:

- When a QRS complex is detected, the detection threshold of the energy signal (ET) is raised in order to avoid detection of T waves:
  $ET \Leftarrow 4 \times \left(0,75 \times ET + 0,5 \times QRS\ peak\right)$ (13)

- 200 ms after detection, the ET is lowered:
  $ET \Leftarrow 0,20 \times ET$ (14)

- If there is no detection for 1 second, the detection threshold is adjusted:
  $ET \Leftarrow 0,5 \times ET$ (15)

When there is a crossing at the detection threshold an event is considered to have happened only after the detector returns to a level bellow the threshold. Therefore, the event will be analyzed by the detector which will classify it as a QRS complex or not.

At each detected event the parameters of RR interval, width and amplitude are calculated. Value ranges considered valid for a QRS complex were determined after several trials. The amplitude mean of the last 8 detected complexes is also calculated.

When an event is detected, its parameters are compared with the range of valid values for a QRS complex.

An event is classified as a QRS complex if: the RR interval is larger than 200 ms, the width is between 16 ms and 500 ms and the amplitude is between 10% and 600% of the amplitude mean from the last 8 QRS complexes.

An example of false detection that is avoided by the secondary detector is shown in Figure 4.

When 1 second goes by without the detection of one QRS complex, the amplitude mean must also be adjusted, in addition to the detection threshold settings, since the signal may have suffered alterations which effectively reduced or increased considerably its amplitude mean. This adjustment is done with the analysis of the amplitude value of the last event. If it is higher than the current mean, the mean value is increased 25%; if it is lower, it is reduced 25%.

One of the best characteristics of this detector is the reduction of false detections in the presence of noise, baseline wandering or movement artifacts. For each of this cases there is most often at least one parameter value off the valid value range for a QRS complex.

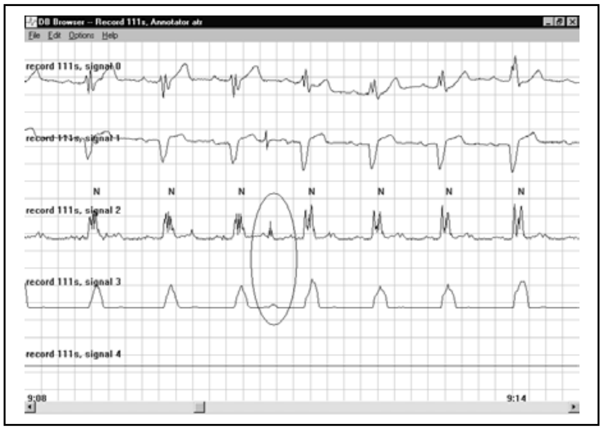

Figure 4 – The circled event is discarded due to its low amplitude.

The detection in two modules takes place in different instants, since the secondary module analyzes the event after the detection of rising and descent crossing, and the main detector waits 180ms after the last crossing. Therefore, the main module keeps results of the last 300ms of the secondary module; when there is a single crossing, the second detector module checks the vector for the existence of a QRS complex, thus guaranteeing a more reliable final result.

3. Results

Testing and analysis of the algorithm were made using the MIT/BIH database. The algorithm performance was evaluated according to sensitivity and positive predictivity, defined respectively as the percentage of QRS complexes detected and the percentage of true QRS complexes among the ones detected. Results were 99,22% to sensitivity and 99,73% to positive predictivity.

Excluding records 108, 200, 201, 203, which presented regions with too much noise and baseline wandering, the results were 99,56% for sensitivity and 99,82% for positive predictivity.

The performance improvement due to the interaction of the two modules can be perceived with the observation of each module performance separately. Results obtained using only the main module were 99,12% for sensitivity and 98,04% for positive predictivity, and results obtained using only the secondary module were 99,08% for sensitivity and 99,62% for positive predictivity.

4. Conclusion

The method described in this article detects cardiac beats in real time, with a maximum delay of 180 ms, and can be used with one or more channels of lead of the electrocardiogram.

This two features, real time detection and use of more than one channel, are important in the introduction of this algorithm in commercial cardiac monitors.

The use of more channels has the advantage of feeding a better input signal, since even if one of the channels has a low quality signal (hindered by noise or by a lead that presents low amplitude) the second channel may contribute positively for the algorithm's input signal. However, if both channels are noisy results may be close to the ones obtained with algorithms used in only one channel.

The combination of two different detection methods improves greatly the results of each method in separate, proving it is possible to add the best features of each method and eliminate their weaknesses.

A possible improvement would be the use of a noise detection algorithm to adjust minimum thresholds of both algorithms proposed here, thus avoiding the detection of false positives even in signals with a low signal/noise rate.

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