Implementation of a simple real-time algorithm for ventricular fibrillation detection in a microcontroller

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Abstract— **This paper describes the implementation of an algorithm for detecting episodes of Ventricular Fibrillation (VF) in a MSP430F6659 microcontroller. Statistical and digital signal processing techniques were used in its design. The validity of its performance was carried out in 3 ECG records from the MIT-BIH Arrhythmia Database, 1 record of the MIT-BIH Malignant Ventricular Arrhythmia Database and in 10 records of the Creighton University Ventricular Tachyarrhythmia Database. These records correspond to patients who suffered of VF episodes during the ECG study. The algorithm was able to detect 100% of VF events and has a time of computation of approximately 100 µs.**

Keywords—**Ventricular Fibrillation, Embedded System, Microcontroller, QRS complex, Signal Processing.**

I. Introduction

Ventricular Fibrillation (VF) is a usually lethal, malignant cardiac arrhythmia characterized by irregular, partial and unsynchronized contractions of the ventricles, with the absence of an effective mechanical activity of the heart. Usually, the VF leads a cardiorespiratory arrest and the death of the patient if immediate treatment (cardiopulmonary resuscitation and electrical defibrillation) is not practiced. From the electrocardiographic point of view, VF is characterized by irregular high-frequency waves (between 300 and 500 BPM) with varying morphology and amplitudes [1]. Fig. 1 shows a portion of an ECG record with this type of cardiac arrhythmia.

Researchers have previously described automated systems for VF detection using various detection methods including time domain [2], frequency-domain [3], wavelet transform [4] and nonlinear analysis [5]. The compressive review of these methods can be found in [6]. In the last years the most important works has focused on the design of algorithms using neural networks [7] [8] [9]. They have the disadvantage that require a training phase and have a high computational cost.

The objective of this work is implementation in a microcontroller of a mathematical algorithm that allows detection of VF after a record of ECG. The algorithm must have a low computational load to enable its use in very low power consumption microcontrollers and allow, in the future, their use in portable medical equipment.

This paper is structured in the following way. Section II explains the algorithm designed for the VF detection. Section III describes the implementation and algorithm validation using simulation software. Section IV shows the implementation of the algorithm in a microcontroller. Finally, section V presents the results and conclusions.

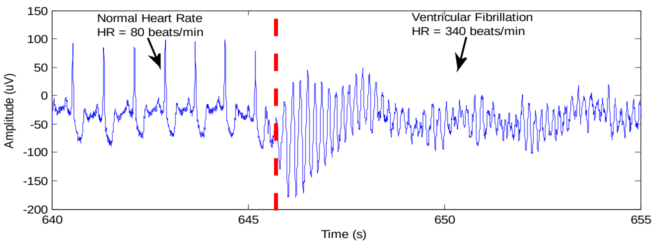

Fig. 1 Ventricular fibrillation (extracted from the Record #426 of MIT-BIH Malignant Ventricular Arrhythmia Database).

II. Ventricular fibrillation detection algorithm

Since VF is directly related to the value of the heart rate, the first task of the algorithm is to estimate its value.

The algorithm implemented in this work is based on QRS complex detector proposed by Pan and Tompkins (1985) [10]. It is suitable for use in continuous monitoring in MCUs since it works in real time and it is simple to design and implement with digital filters, which also have a low computational cost.

Fig. 2 shows the block diagram of the algorithm designed. The ECG signal enters a band-pass filter, whose objectives are to eliminate baseline variations, attenuate the low frequencies of P and T waves, reduce the lines interference of 50/60 Hz and the high-frequency noise, and conserve bandwidth where most of the energy of the QRS complex is concentrated.

The output signal of the band-pass filter is led to a differentiator filter, which allows the enhancement of the high frequencies of the QRS complex by attenuating the residual low frequency components of P or T waves that may still be present.

As shown in Fig. 2, the output signal of the differentiator filter is then led to a non-linear transformation block for the purpose of obtaining high positive values in the signal segments corresponding to the QRS complexes. To do this, the quadratic value to the output of the differentiator filter is calculated.

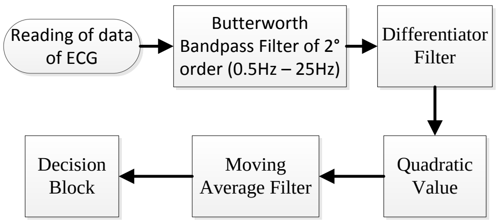

Fig. 2 Block diagram of the algorithm designed.

To smooth the quadratic value, and obtain a single peak that represents the QRS complex, a moving average filter is used, which averages a certain number of consecutive samples; the greater that number, the smoother the obtained signal will be.

The decision block determines the presence of the QRS complex. It is based on a threshold detector, which is modified according to the historical amplitude of the signal. This is an important stage, on which the efficiency of the designed algorithm depends. Specifically, the threshold value depends on the median of a certain amount of previously detected peak values of heartbeats. In case that heart beats are not being detected, the maximum value of a temporary window with a certain length and then the median is calculated taking into account this new value.

The elapsed time between two consecutive QRS complexes is finally measured. If this is less than a limit previously set by a medical professional (usually 300 BPM), the algorithm marks the presence of a VF episode.

III. Implementation and Validation Using Simulation Software

For the design of the band-pass digital filter an IIR-type with a frequency response of the Butterworth type was used. The calculation of the coefficients was carried out with the tools provided MatLab® software, considering the sampling frequency of 250 Hz of ECG records used to perform validation.

Once the filter was designed, its implementation was carried out using the algorithm of the equations in differences (1), as well as the rest of the blocks (2) (3) (4). The latter ones, because of their simplicity, did not require any calculation of coefficients.

Band-pass filter

$$\begin{aligned} BW(n) &= 0.034\times ECG(n) - 0.0696\times ECG(n-2) \\ &+0.0348\times ECG(n-4) + 3.4007\times BW(n-1) \\ &-4.3496\times BW(n-2) + 2.4955\times BW(n-3) \\ &-0.5465\times BW(n-4) \end{aligned} \tag{1}$$

Differentiator Filter

$$DR(n) = Fs\times\left[BW(n) - BW(n-1)\right] \tag{2}$$

Quadratic value

$$Q(n) = DR(n)\times DR(n) \tag{3}$$

Moving Average Filter

$$Av(n) = \frac{1}{32}\times\sum_{i=0}^{31} Q(n-i) \tag{4}$$

Fig. 3 shows the output of each of the blocks.

Fig. 4 is a set of pulses typical of the output of moving average filter; it was used to stablish the decision block algorithm. The flowchart of the latter is show in Fig. 5.

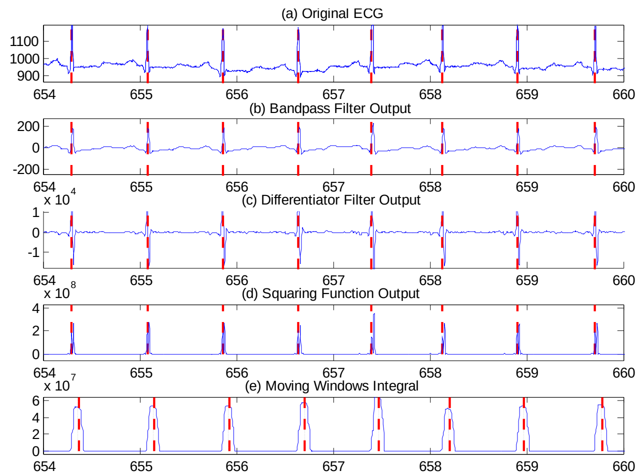

Fig. 3 (a) Original ECG; (b) Bandpass filter output; (c) Differentiator filter output; (d) Quadratic output and, (e) Moving average filter.

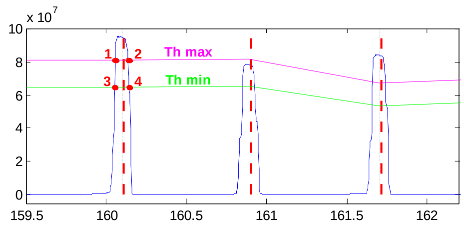

Fig. 4 QRS detection thresholds

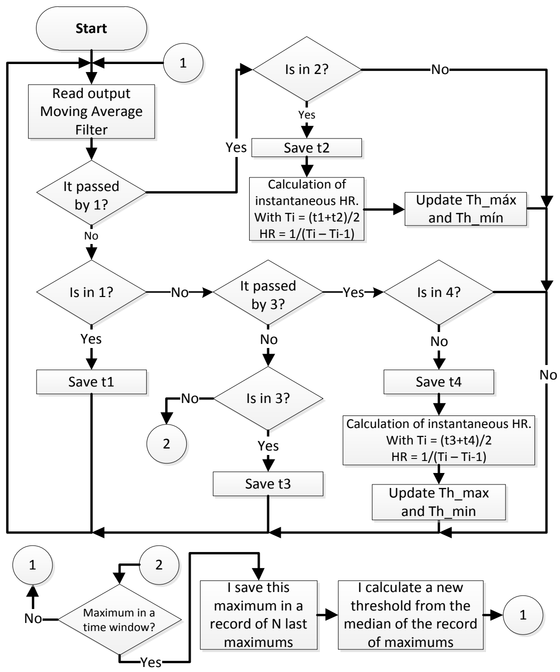

Fig. 5 QRS decision block flowchart

Fourteen ECG records of patients who suffered episodes of VF were especially selected for validation of algorithm perfomance. Specifically, 3 records from the MIT-BIH Arrhythmia Database were selected (1 male, age 69; 2 female, age 79±5) [11], 1 from the MIT-BIH Malignant Ventricular Arrhythmia Database (no available patient data) [12] and 10 from the Creighton University Ventricular Tachyarrhythmia Database (no available patient data) [13], all available free of charge in the Physionet website. In Table 1 we can observe the performance of the QRS complexes detection algorithm in ECG records from the MIT-BIH AD database, based on the following indexes:

Table 1. Performance of the QRS complexes detection algorithm in different ECG database records from AD MIT-BIH

Database	Rec. #	Beats	TP	FP	FN	FD
MIT-BIH AD	100	2273	2269	2	3	5
MIT-BIH AD	101	1874	1862	2	10	12
MIT-BIH AD	102	2192	2177	0	3	3
Total		6339	6308	4	16	20
Percentage of correct detections		99,51%				
Percentage of false detections		0,31%				

The initials correspond to: TP = True Positive; FP = False Positive; FN = False Negative; FD = False Detections.

The Table 2 shows the records that have been used to validate the VF detector, the number of VF events indicated by physicians in each record, the quantities of correct (TP) and incorrect (FN) detections.

This work considers a VP when the algorithm detect event in less than 5 seconds since VF event start. It can be noted that 100% of events VF were detect in the 11 analyzed ECG records.

Table 2. Result of ventricular fibrillation algorithm

Database	Record #	VF Events	TP	FP	FN
CUDB	CU01	1	1	0	0
CUDB	CU03	1	1	0	0
CUDB	CU04	4	4	0	0
CUDB	CU06	1	1	0	0
CUDB	CU07	1	1	0	0
CUDB	CU08	1	1	0	0
CUDB	CU10	1	1	0	0
CUDB	CU11	1	1	0	0
CUDB	CU12	1	1	0	0
CUDB	CU13	1	1	0	0
VFDB	426	2	2	0	0

IV. IMPLEMENTATION IN A MICROCONTROLLER

A. Hardware

The designed algorithm was implemented in a microcontroller MSP430F6659 [14] of Texas Instruments®, which has a CPU of 16 bits and 32 x 32-bits multiplier in hardware (MPY). The latter allows the optimization of processing times. It is also recommended for portable applications, due to its low consumption of energy. An ADS1298RECGFE-PDK [15] kit was used for the acquisition of ECG signal.

To validate the performance of the hardware model a multiparameter simulator PS420 of the firm Fluke was used, calibrated at the time of the tests.

B. Firmware

The software that runs the MCU was implemented in a C programming language, using the development environment CCS (Code Composer Studio) provided by the manufacturer of the chip.

The MCU reads the data from the interface of acquisition through a SPI (Serial Peripheral Interface) communication, at a speed of 250 samples per second and a resolution of 24 bits.

The most critical action of conditioning, from the acquired ECG signal, that the MCU must perform is the band-pass filtering. The type IIR filter selected for this job, presents as a disadvantage the possibility of being unstable if it is not properly designed. In addition, once designed, the coefficients cannot be truncated arbitrarily. If this is done, the stability of the filter must be checked.

As the MPY of selected MCU does not allow operations in a floating point, and doing it by software would require more computing time, the filter coefficients were truncated and were transformed into integers. This solution was chosen because the filter is of a 2nd order and there are not many coefficients to adjust.

To optimize computational time in the operations of calculating division, byte rotation technique was used. While the result is not accurate, it is a good approximation. An example of this is shown below:

```
// for C = 0.8 * Ti
C= (Ti >> 2) + (Ti >> 4) // 0.75 * Ti
```

To calculate the median value, an array of data with the last ten maximum values of the output of the moving average filter was used and through the bubble sort method. Then, the median is the data that is halfway between the major and minor. The calculation of the median is performed by comparing the data and it does not involve a significant computational cost, considering that the amount of data is minimum [16].

To transmit the results of the implemented code in the MCU to a PC, an asynchronous serial communication was used, and for the visualization of the data, the Tera-Term software.

V. Conclusions

In this work we have designed an algorithm that is capable of estimating instant heart rate and detect a VF event.

The algorithm was implemented in a microcontroller MSP430F6659, optimizing the code to reduce the processing time. The latter is in the order of 100µs, at a rate of 8 MHz processor and was obtained through a logic analyzer.

The performance of the algorithm has been measured with 14 real ECG records from 3 validated databases. In three of these records from the MIT-BIH Arrhythmia Database, 6339 beats were analyzed and 99.51% of correct detections were achieved. The algorithm was also tested with the record # 426 MIT-BIH Malignant Ventricular Arrhythmia Database, and 10 records from the Creighton University Ventricular Tachyarrhythmia, and comparing the obtained results with medical indications in the records, it was proved that the algorithm correctly detected the all VF events. Finally, it should be noted that 100% of events VF were detected in the 11 analyzed ECG records as is shown in Table 2.

Its advantage that no need to be trained like required other methods [7][8][9] that use neural networks. Also, it have low load computational. Due to its simplicity, it is noteworthy that the proposed algorithm could be implemented in portable equipment of ambulatory cardiac monitoring.

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