AN ALGORITHM FOR VENTRICULAR FIBRILLATION DETECTION

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Summary: *Ventricular fibrillation detection from the external electrocardiogram (ECG) is a difficult task, involving high decision making responsibility. An algorithm is proposed, based on assessment of the signal shape and frequency. It reaches 90% sensitivity and has a specificity of 92%, as evaluated using the widely recognized ECG signal databases of the American Heart Association and the Massachusetts Institute of Technology.*

Introduction

Reliable and correct external electrocardiogram (ECG) signal analysis is of crucial importance for further development of automatic external defibrillators (AED) and their use by inexperienced helpers or attendants. They should provide accuracy comparable to this of trained medical personnel.

Many software methods for ventricular fibrillation (VF) detection have been proposed. In the time-domain, the most well-known are the threshold crossing intervals (TCI) (Thakor *et al* 1990) and auto-correlation function (ACF) methods (Chen *et al.*, 1987).

The first method uses setting of a threshold at 20% of the maximum value in 1s time-segments. The intervals between two consecutive crossings of the threshold are measured. Then the mean TCI is calculated. If its value exceeds 400ms, the segment is classified as non-VF. Otherwise a complicated testing process is used to separately classify VF from ventricular tachycardia (VT). The short-term ACF separates VF from normal heart activity by a periodicity assessment.

In the frequency-domain two methods are favoured – the spectrum analysis (Barro *et al.*, 1989) and the VF-filter (Kuo and Dillman 1978) recently assessed by CLAYTON *et al.*, (1993).

The VF-filter technique corresponds to a narrow band-stop filter applied to the signal, with central frequency equivalent to the mean signal frequency. Its output is the VF-filter leakage. The spectrum algorithm calculates some signal spectrum parameters and determines correspondingly the rhythm type.

Method and algorithm

Recordings from the AHA and MIT databases were processed by means of the software package MATLAB. They were evaluated by each of the above mentioned techniques (in a previous study) and by the method described in this paper.

All methods considered here depend on comparison between measured values and thresholds, characteristic of VF. To be representative and reliable, these threshold values must be computed from a large population (Clayton *et al.*, 1993).

The method proposed in this paper is based on the signal shape and frequency. First a preprocessing was applied on all signals using: i) a notch filter to virtually eliminate powerline interference, ii) a high-pass filter with cutoff frequency at 1 Hz to suppress residual baseline drift, and iii) a second-order low-pass Butterworth filter with a cutoff frequency at 30 Hz to reduce muscle noise, following the approach of Thakor *et al.* (1990).

The algorithm is composed of several steps:

1. The maximum and minimum signal values in the examined period are found – Smax and Smin (S is the signal string) and their absolute values are compared to each other.
2. Thresholds marked by T at ±50% of the higher absolute value are set.
3. New signal string Sn is generated. If (-T)<Si<(T) Sn=0; else if (–T)>Si Sn=Si+T; else if T<Si Sn=Si-T.
4. The positive and negative peaks which are not closer than 200ms (50 samples) keep their values. All other samples become 0. Thus the Sm string is obtained – Fig. 1 (2nd trace) to 8 (2nd trace).
5. The positive and negative peaks in Sm are counted. Only one polarity peaks is taken whose number is greater– Fig. 1 (3rd trace) to 8 (3rd trace). After that the number of the intervals is calculated - int_num=(peaks number – 1)
6. If int_num<(3*dur-1), where dur is the duration of the assessed period and int_num>0:
   - The duration of each interval is calculated.
   - The mean interval value is computed.
   - The number of intervals in the range (0.7*mean interval ÷ 1.3*mean interval) are counted and if they are less than int_num-2 the rhythm is identified as VF. Otherwise it is not VF.
7. If int_num ≥ (3*dur-1) the rhythm is eighter VF or shockable VT (above 180 beats/min).
8. If there is only one peak found (no interval can be determined), the algorithm generates an "undecided" output.

Results and discussion

ECG signals are considered non-shockable if they are of sinus rhythm or if the heart frequency is <180min^{-1}, regardless of rhythm type. 77 non-shockable 8s segments (among which 6 VT) and 90 shockable 8s segments were assessed. The specificity and sensitivity obtained by the algorithm are respectively 92% and 90%. The VF-filter yielded 91% specificity and 94% sensitivity with the same data segments. It came out to be of the best performance among the four techniques described above.

Parts of ECG recordings containing sinus rhythm (SR), VT and VF are shown in eight figures. The first four cases are correctly identified as shockable or non-shockable rhythms. The next three are erroneously recognized. The last case is "undecided".

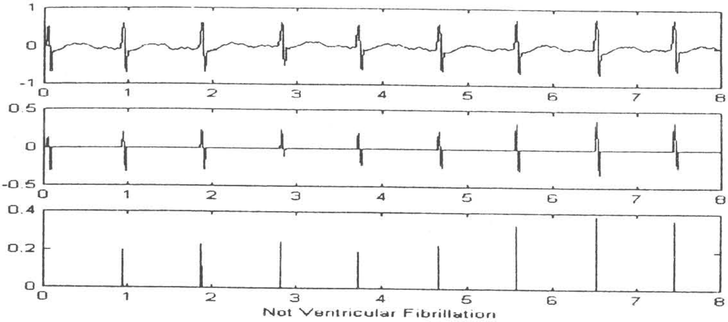

Fig.1

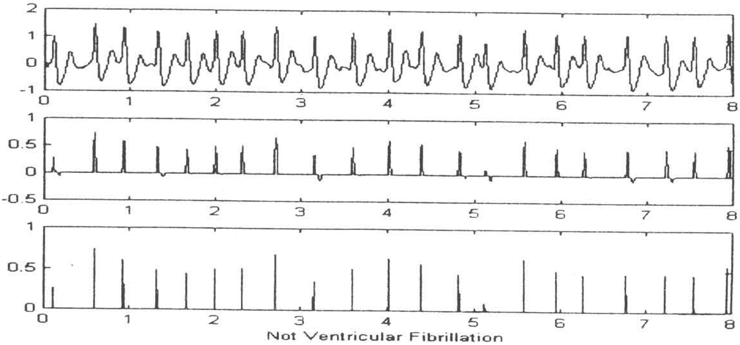

Fig. 2

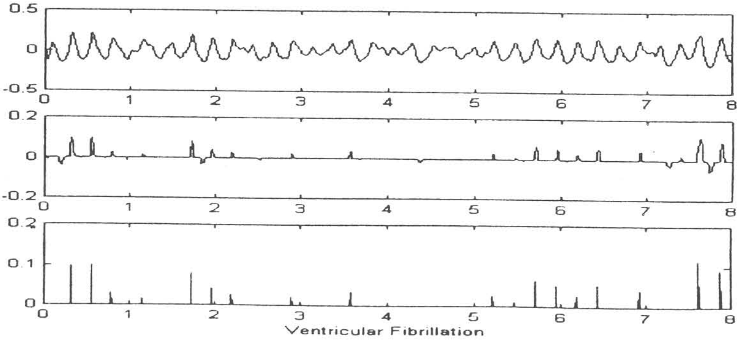

Fig. 3

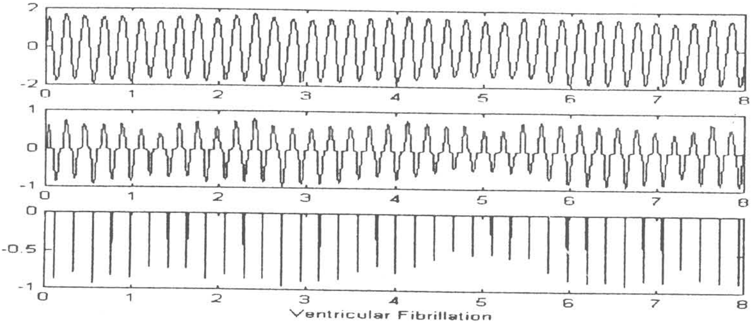

Fig. 4

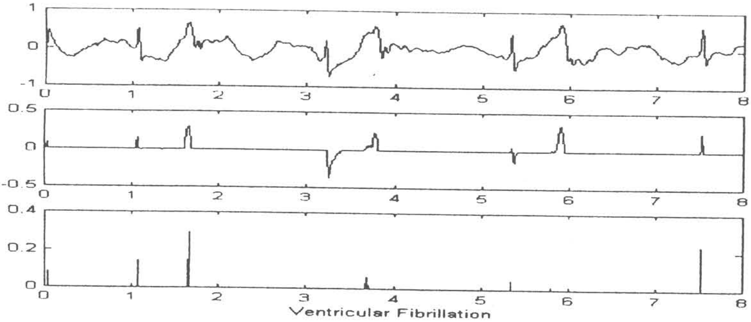

Fig. 5

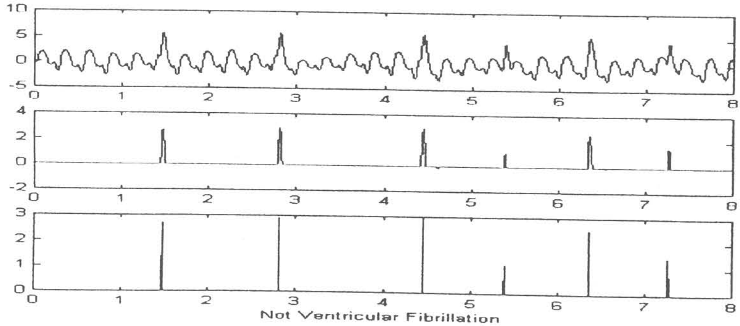

Fig.6

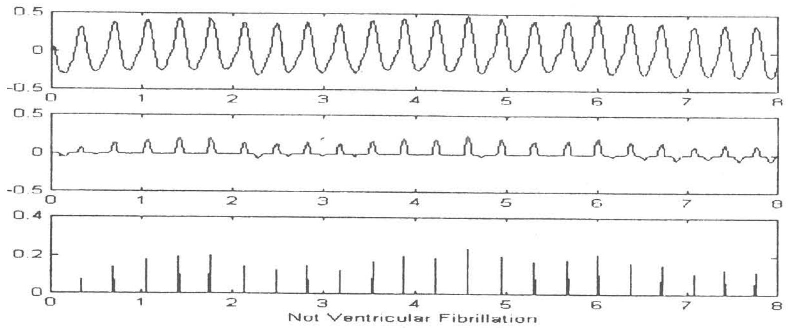

Fig. 7

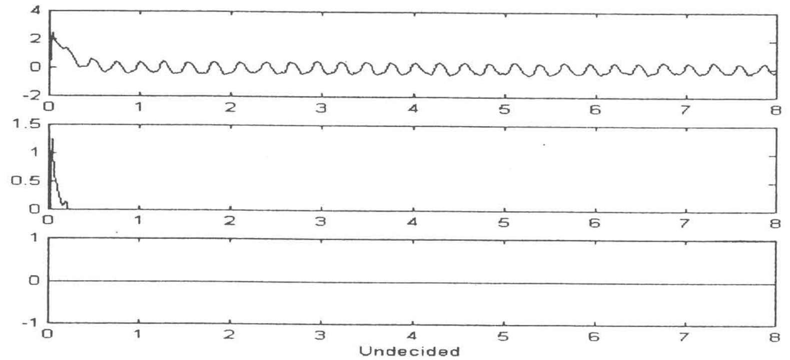

Fig. 8

Conclusion

AED are becoming of increasingly wide use and the requirements for accurate and reliable separation between shockable and non-shokable rhythms will continue to build up. The proposed algorithm can be further improved in order to yield still better sensitivity and specificity.

References

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