AN ALGORITHM FOR THE DETECTION AND CLASSIFICATION OF ATRIAL FIBRILLATION FROM INTRA-ATRIAL ELECTROGRAMS

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Abstract

Reliable detection of atrial fibrillation from intra-atrial signals is a central task in the development of implantable atrial defibrillators. Atrial rate (AR) and the amplitude probability density function (APDF) are normally used as parameters for discriminating between normal sinus rhythm (NSR) and atrial fibrillation (AF). We hereby propose a new peak-detection scheme based on threshold crossing. The main features of the algorithm are: prefiltering of the atrial electrograms magnitude with FIR band-pass filter and the combined use of root-mean square and of peak amplitude for the setting of an adjustable threshold. A blanking period and an exponential time decay of the threshold have also been introduced to increase peak-detection robustness. The classification of the AF rhythm, according to Wells’ classes, has been obtained by the combined use of the coefficient of variation of AR (CV) and of APDF values. To find the optimal settings and to test the performance and robustness of the peak-detection algorithm we used data from intra-atrial recordings from chronic AF patients and from subjects with electrically induced AF. On the basis of these settings we implemented the classification scheme. The peak-detector revealed a 98.75% sensitivity, 98.33% specificity and 97.07% overall accuracy. Although CV and APDF satisfactorily matched the Wells’ criteria, we feel that this should be further confirmed by a larger statistical analysis.

Introduction

The robust and accurate detection of atrial fibrillation (AF) from intra-atrial electrograms is of primary importance in the design of implantable devices; moreover the classification of AF in different types, as suggested by Wells [1], could be used for the choice of shock delivery strategies. Algorithms for AF detection are mainly based on atrial rate (AR), amplitude probability density function (APDF), power spectrum density analysis (PSD) and correlation waveform analysis (CWA) [2][3]. Several authors have compared the different approaches and algorithms in terms of sensitivity and specificity of AF detection. Nevertheless, a further classification of AF according to Wells’ criteria has been seldom addressed. Wells’ classification divides AF in 4 distinct types: Type I (AF1) is characterised by discrete beat-to-beat atrial electrogram complexes of variable morphology and cycle-length, separate by an isoelectric baseline free of perturbation. Type II (AF2) is similar to type one, but the baseline is not isoelectric and presents various perturbations [1]. In Type III (AF3) discrete intervals and isoelectric segments are no longer detectable. Type IV (AF4) is characterised by atrial electrograms of Type III, alternating periods of electrograms consistent with Type I and/or II.

We based our classification of AF on two variables: APDF and the coefficient of variation of the atrial rate (CV). APDF encodes information about the presence of a baseline in the electrograms, while CV indicates whether electrograms are characterised by detectable complexes.

Aim of this study was to design a new algorithm for the discrimination between normal sinus rhythm (NSR) and AF from intra-atrial electrograms. The proposed procedure uses an adjustable threshold peak-detection scheme. In addition an AF classification algorithm is introduced: this approach combines the information furnished by atrial rate and by the amplitude probability density function for rhythm classification according to Wells’ criteria.

Methods and Materials

The study was performed in 25 informed subjects: 10 chronic AF (CAF) and 13 sinus rhythm + electrically induced AF (IAF). Intra-atrial electrograms were real-time digitised (1000 samples/s, 16 bit resolution), during haemodynamic cardiac catheterisation. Each recording was then visually inspected by an expert cardiologist to determine the true atrial depolarisations and to classify the rhythms as follows: normal sinus rhythm (NSR) and atrial fibrillation of types I, II, III and IV, according to Wells’ criteria. We considered segments of at least 2 seconds of the same rhythm (i.e. NSR, AF1, AF2, AF3). Segments including AF4, or any other arrhythmia or a change from NSR to AF or vice-versa were disregarded. The overall procedure contains two main subsections. The first is the depolarisation
detection algorithm, based upon an adjustable threshold crossing, the second the classification scheme of the rhythm which uses the combined CV + APDF information.

**Peak detection.** The block diagram of the peak-detection algorithm is shown in fig.1. The scheme contains a FIR band-pass filter (40-250Hz, order 40, Kaiser window) to remove baseline shift and high frequency noise. The absolute value of the output of the band-pass filter was then low-pass filtered (FIR, 0-30Hz, order 40, Kaiser window). This process extracts a time-varying waveform proportional to the amplitude of the high frequency components of the original atrial electrograms [5]. The trigger has an adaptable threshold with exponential time decay to allow for variability in waveform amplitude. Moreover, a blanking period has been imposed to prevent multiple detection on single depolarisation.

The equation for the adjustable threshold is:

\[
th_i(t) = (M_i - \sigma_T) e^{-\left((M_i - \sigma_T)(t - t_i - \text{blank})\right) / \tau} + \sigma_T \quad \text{for } t > t_i - \text{blank}
\]

where:

\[
M_i = \text{amplitude of the previously detected peak}
\]

\[
\sigma_T = k_{\text{RMST}} + \text{RMS}_T + \sigma_{\text{abs}}
\]

\[
k_{\text{RMST}} = \text{fractional constant (ranging between 0 - 1)}
\]

\[
\text{RMS}_T = \text{root mean square of previous } T \text{ seconds of the signal}
\]

\[
T = \text{time window RMS estimation}
\]

\[
\sigma_{\text{abs}} = \text{absolute threshold}
\]

\[
t_i = \text{occurrence time of the } i^{th} \text{ peak}
\]

\[
\tau = \text{time constant of the exponential decay}
\]

\[
\text{blank} = \text{blanking period}
\]

The threshold \(th_i\) is thus function of the amplitude of the previously detected peak and of the RMS of the last \(T\) seconds of the signal. Due to the high variability of the signal between NSR and AF, the amplitude of the previously detected peak has been used as reference for the new threshold value. The exponential decay is controlled by the time constant \(\tau\) and by the difference \((M_i - \sigma_T)\). This difference accounts for the large variability in the waveform amplitude: the larger the peak, the steeper the exponential decay, which modulates the sensitivity of the trigger. The starting time for the exponential decay is set at the end of the blanking period. Finally, an absolute minimum threshold \(\sigma_{\text{abs}}\) has been used to keep the threshold over the noise floor.

Of the 10 (CAF)+13 (IAF) patients considered, data from 2+3 were used to define the range of variability of the peak-detector parameters. Then, we evaluated the algorithm performance and the optimal parameters. To this purpose, from each patient, we selected a maximum of 100 depolarisations of any type, thus obtaining a total number of 904 NSR depolarisations, 926 IAF depolarisations and 800 CAF depolarisations, previously detected and classified by hand by an expert cardiologist.

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**fig.1** Block diagram of the peak-detector.
Rhythm classification. For discriminating between NSR and not-NSR (i.e. AF of any type) we used the average AR (AR\textsubscript{aver}) over the last 5 peaks detected. To increase robustness, those values falling outside the ±2 standard deviation were not included in the computation of the AR\textsubscript{aver}. On the basis of literature data, decision threshold was set at 300ms [2].

AF classification in types I, II, and III was based on a combined approach which uses both the coefficient of variation (CV) of AR and the amplitude probability density function (APDF). CV was defined as the standard deviation of the last 5 depolarisation intervals divided by the AR\textsubscript{aver}. The APDF function was calculated from the raw electrograms, as the probability of occurrence in the maximum populated bin within a proper interval centred on the baseline. We assumed that the pair (APDF, CV) defines a two-dimensional set in which Wells' classification is projected as depicted in fig.2. APDF is supposed to estimate the number of points of the signal laying on the baselines; CV accounts for the presence of clearly detectable peaks.

![fig.2 Regions representing Wells' classes projected on the two-dimensional space APDF-CV.](image)

We also provided a non-classified class (NC) to account for changes from NSR to AF and/or among AF types and for borderline conditions. Abrupt changes in electrogram type are detected by high CV values and/or by beat-to-beat differences in CV (ΔCV). In our preliminary data, we set the threshold levels empirically. The classification scheme is shown in fig.3.

![fig.3 Flow chart of the classification section.](image)

**Results**

Fig.4 shows a typical recording from one subject during NSR and IAF. In this recording some typical electrogram patterns which can affect the accuracy of the algorithm can be observed. This patterns include:

A. presence of far-field ventricular depolarisation;
B. short-term variability of atrial depolarisation waveforms;
C. abrupt changes in AF type, with atrial depolarisation and baseline difficult to identify.
The overall results of the peak-detection algorithm are summarised in table 1 and 2. T was set at 3 sec, and \( \sigma_{\text{abs}} \) at 4 LSB (least significant bit). The optimum parameter setting (i.e. the set which balances algorithm performance under the three conditions), which also has been used the rhythm classification, was: \( K_{\text{rms}}=0.2 \), \( \tau=80 \) and \( \text{blank}=50 \). For this set, of the total 2630 depolarisations studied, which included any type of rhythm (i.e. NSR, Type I CAF and IAF electrograms), the triggering algorithm missed (false negative) 33 (sensitivity 98.75\%) and gave 44 false positive (specificity 98.33\%), with an overall accuracy of 97.07\%. The performance in AF yielded 20 false negative (sensitivity 97.50\%) and 2 false positive (specificity 99.75\%) for CAF, and 13 false negative (sensitivity 98.60\%) and 8 false positive (specificity 99.14\%) for IAF. In NSR we collected 0 false negative and 34 false positive, thus achieving a 100\% sensitivity and 96.24\% specificity.

**table 1. Results of the peak-detector.**

<table>
<thead>
<tr>
<th>Parameters</th>
<th>NSR</th>
<th>IAF</th>
<th>CAF</th>
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<tbody>
<tr>
<td>80</td>
<td>65</td>
<td>0.2</td>
<td>904</td>
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<td>60</td>
<td>65</td>
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</tr>
<tr>
<td>80</td>
<td>65</td>
<td>0.3</td>
<td>904</td>
</tr>
</tbody>
</table>

N=number of depolarisations; FP=false positive; FN=false negative; NSR=normal sinus rhythm; IAF=induced atrial fibrillation; CAF=chronic atrial fibrillation.

**table 2. Overall performance of the peak-detector.**

<table>
<thead>
<tr>
<th>Parameters</th>
<th>NRS+IAF+CAF</th>
</tr>
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<tbody>
<tr>
<td>( \tau )</td>
<td>Bl.</td>
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<tr>
<td>80</td>
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Sensitivity, specificity and accuracy in percentage values.
Rhythm classification. Fig. 5 shows an example of AF classification. AR_{ravel}, ΔCV, CV and APDF for AF3 (panel A), AF2 (panel B) and AF1 (panel C) are listed.

![AF classification diagram](image)

**Fig. 5** Example of AF classification using CV and APDF: the original signal (upper trace) and its low-pass filtering (lower trace) for AF3 (panel A), AF2 (panel B), AF3 (panel C)
The upper trace of each panel shows the original electrogram, the lower trace the result of the filtering process for the peak-detection. In addition the threshold level with the exponential decay is plotted. Each detected depolarisation is marked and labelled according to the CV-APDF pair. Although CV progressively increases from AF1 to AF3 and APDF decreased from AF1 to AF3, a reliable classification is achieved only with a combined use. APDF best discriminated AF1, while overlap exists between AF2 and AF3. Conversely CV allowed to discriminate AF3, while overlap region exist for AF1 and AF2.

Discussion and Conclusions

As previously shown, the peak-detection algorithm performed well during atrial fibrillation. The major number of missed triggerings (false negative) occurred in electrograms in which pairs of large-amplitude waveforms were followed by a small amplitude one. The few number of false positive were due to far-field ventricular depolarisation. During sinus rhythm the incidence of false positive triggering was unsatisfactorily high in those patients (2/10) with electrograms corrupted by relatively large ventricular complexes. Since a reduction of the sensitivity of the peak-detector would result in an unacceptable increase in missed detections during AF, the use of a blanking period triggered by a ventricular signal is strongly suggested. Although false positive detection of ventricular complexes during NSR does not result in a misclassification of NSR under normal condition, this can still happen during atrial tachycardia.

The missed beats in AF never affected the correct discrimination between AF and NSR, for they occurred randomly and were usually disregarded in the estimation of the AR aver. As a result AR aver is satisfactorily robust for discriminating between NSR and non-NSR (i.e. AF of any type).

The combined use of CV and APDF seems to allow for a good discrimination of the different types of AF. APDF reliably estimated the number of point laying on the baseline in the electrograms; low CV values occurred in those electrograms in which atrial complexes were clearly detectable. In conclusion, although the combined use of CV and APDF seems to match Wells’ criteria, we feel that this should be confirmed by a larger statistical analysis.

References