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Automatic detection of premature atrial contractions in the electrocardiogram

Vessela T. Krasteva, Irena I. Jekova, Ivaylo I. Christov

The analysis of the electrocardiographic (ECG) signals, especially the QRS complex as the most characteristic wave, is a widely accepted approach to study and to classify cardiac dysfunctions. The detection of atrial premature beats is considered clinically important, since it is a sign for disturbance in the depolarization process preceding in many cases the appearance of supraventricular tachycardia, postoperative atrial fibrillation and paroxysmal atrial fibrillation. The developed algorithm for detection of atrial premature beats analyses two channels of ECG signals. In general, the method consists of preprocessing filtration, heartbeat detection and heartbeat classification by estimation of the interbeat differences of the RR intervals and the QRS morphological descriptors.The testing of the algorithm with the publicly available MIT-BIH arrhythmia database presented a relatively high accuracy with sensitivity of 92.2 % and specificity of 96 %, which is higher than the accuracy of reported algorithms. We proved that the studied parameters are a reliable set for detection of atrial premature heartbeats in all typical arrhythmias. The developed algorithm is suitable to be implemented as a branch of a more complicated method for heartbeat classification. It is highly effective for detection of atrial premature beats, but an additional approach for discrimination between normal and premature ventricular beats is advisable for extended diagnostics.

Автоматично разпознаване на предсърдни екстрасистоли в електрокардиограмата (Весела Ц. Кръстева, Ирена И. Жекова, Ивайло И. Христов). *Най-често прилаганият метод за изследване и диагностика на сърдечни заболявания е анализът на електрокардиограмата (ЕКГ), особено на морфологията и интервалите между вълните, които съответстват на камерните контракции - QRS комплексите. Като част от този анализ, важно приложение има детекцията на преждевременни предсърдни контракции за предсказване на надкамерна тахикардия и следоперативна или пароксизмална предсърдна фибрилация. Разработен е алгоритъм за автоматичен анализ на двуканална ЕКГ за детекция на преждевременни предсърдни контракции. Методът се прилага в квази-реално време и включва предварителна филтрация на ЕКГ сигнала, QRS детекция и класификация на QRS комплексите, чрез изчисляване на няколко параметъра, оценяващи промяната на формата на комплексите и интервалите между тях. Алгоритъмът е тестван с международно признатата MIT-BIH база данни, съдържаща разнообразни ЕКГ аритмии. Получени са стойности за чувствителност Se=92.2 % и специфичност Sp=96 %, които са относително високи, сравнени с тези на други публикувани алгоритми. Предложеният алгоритъм би могъл да бъде част от по-сложен метод за класификация на сърдечните контракции. Той има висока ефективност при детекция на преждевременни предсърдни контракции, но за по-разширена диагностика се препоръчва неговата комбинация с друг високоефективен метод, специализиран за разпознаване на преждевременни камерни контракции.*

Introduction

The electrocardiographic (ECG) signal represents the electrical activity of the heart. The analysis of this signal, especially the QRS complex as the most characteristic wave in ECG, is the widely accepted approach to study and to classify cardiac dysfunctions. The normal ventricular contractions are represented with QRS complexes with a relatively constant waveform and a relatively regular appearance of each heartbeat. There is only a slow natural variation of the interbeat RR intervals, corresponding to the respiratory activity. Any cardiac dysfunction associated with excitation from ectopic centers anywhere in the myocardium leads to premature contractions (with supraventricular (atrial) or ventricular origin), which alter the RR interval duration and/or the QRS waveform. Usually, the premature atrial contractions have the same conduction path through the ventricles thus producing a normal QRS complex only with different coupling RR intervals. In contrast, the most specific feature of the premature ventricular contractions is the wide QRS with bizarre waveform, due to the abnormal change of both the conduction path and velocity during propagation of the activation front through the ventricles [1].

The premature beat (PB) itself does not cause symptoms but the occurrence of multiple single PBs is considered clinically important, since it is a sign for disturbance in the depolarization process preceding in many cases the appearance of malignant cardiac arrhythmia. The problem of ventricular PBs detection is widely discussed in the literature, since they disturb the blood pumping function of the ventricles and their appearance during the vulnerable period increase the risk for initiation of ventricular tachycardia, ventricular flutter/fibrillation leading to sudden cardiac death. However, it seems that the problem for detection of atrial PBs has been left aside, although they could precede initiation of supraventricular tachycardia and might be used as a predictor for the occurrence of arrhythmias like postoperative atrial fibrillation [2] and paroxysmal atrial fibrillation [3], [4]. Moreover, the reported algorithms for detection of atrial PBs are of limited accuracy, e.g. Tsipouras et al [5] use RR interval durations and declare sensitivity of 75.8 %, Chazal et al [6] apply morphology and heartbeat interval features and obtain sensitivity of 75.9 %, Lagerholm et al [7] implement Hermit functions and self-organizing maps and report sensitivity of about 87 %. Recently, Christov [8] proposed an algorithm, which in the first step detects the ventricular PBs by analysis of the QRS shape variation, followed by a step for discrimination between the normal beats and the atrial PBs by estimation of RR intervals. Since the used testing database has no annotations for atrial PBs, no accuracy for their recognition was reported in the article.

This study is aimed at development of an algorithm for detection of atrial PBs involving into analysis the most effective QRS morphological parameters, which were used in previous studies of the authors for automatic classification of ventricular PBs [9], [10]. Here, we introduce the QRS waveform morphology differences and a relative estimation technique, which allows application of uniform criteria for all patients, regardless of the QRS complexes morphology, specific for each individual. Moreover, the atrial PBs, which appear for defined atrial arrhythmias are detected by setting of specific rules, based on interbeat differences of both the QRS morphology and the RR intervals.

Method

The developed algorithm for detection of atrial PBs analyses two ECG leads. In general, the method consists of three processing stages, including preprocessing filtration, heartbeat features measurement and heartbeat classification. All calculation procedures allow the quasi real-time operation of the algorithm. The general flowchart of the algorithm is presented in fig. 1.

Preprocessing filtration

The applied digital filtering prevents against power-line interference, tremor noise and base-line drift distortions of the input ECG signal that impede the accurate measurement and classification of the heartbeats. The preprocessing filtration is aimed at:

- power-line interference elimination with notch filter, which is implemented by moving averaging of samples in one period of the interference;
- suppression of the tremor noise with a low-pass filter, realized by moving averaging of samples in 30 ms time-interval, thus having a first zero at about 35 Hz;
- drift suppression with a high-pass recursive filter [11] with cut-off frequency of 2.2 Hz.

Heartbeat features measurement

The QRS detection is the initial procedure that locates the appearance of ventricular contractions. Many algorithms for QRS detection are known from the literature, the newest ones being [12] and [13]. We applied [13] because of the reported better accuracy and simultaneous performance with two ECG channels. The detected R-wave peak is used as a reference for the QRS pattern recognition technique [14], which identifies the onset and the offset of the QRS complex by parallel analysis of the two ECG leads. An example of ECG signal (Lead 1 and Lead 2) containing 3 types of heartbeats (normal beats, atrial PB and ventricular PB) with identified R-wave peaks and patterns onset and offset is illustrated in fig. 2(a). Then, several heartbeat features, including both the interbeat RR intervals differences and the QRS waveform morphology differences are measured.

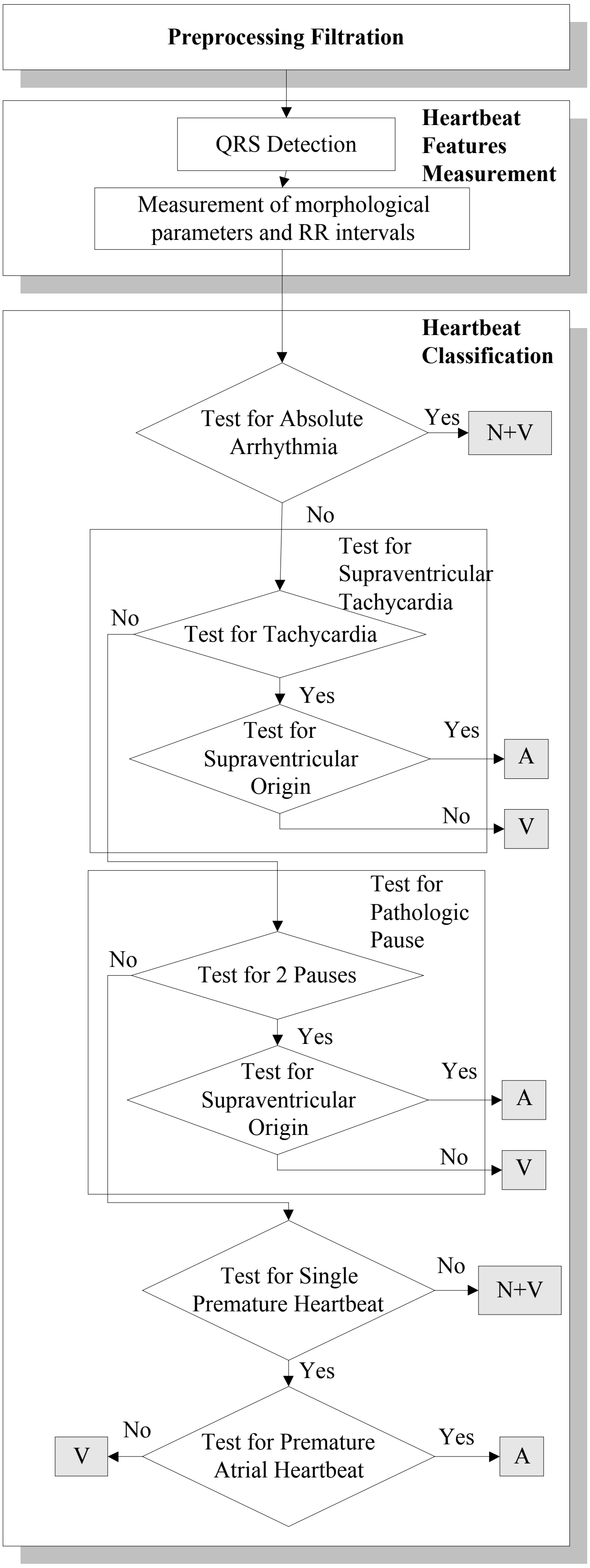

Fig.1. General flowchart of the algorithm
A - Decision for atrial premature beat;
V - Decision for ventricular premature beat;
N - Decision for normal beat.

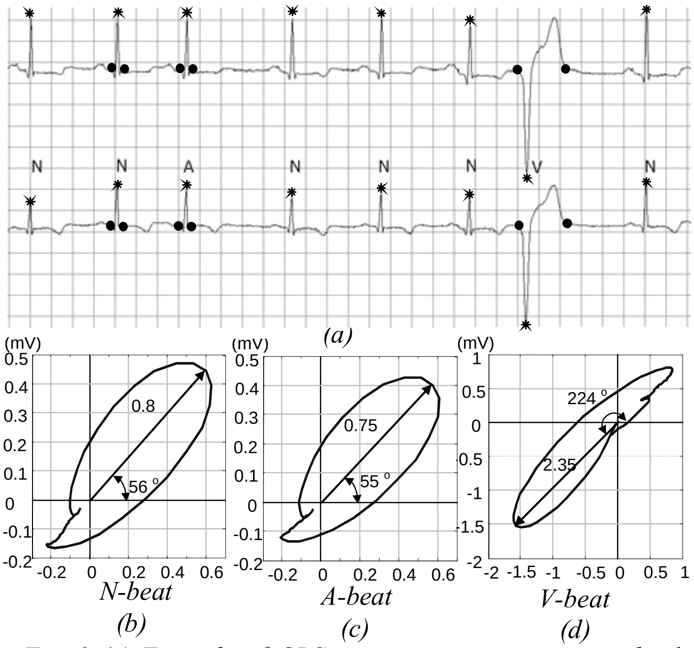

Fig. 2 (a) Example of QRS pattern recognition in two-lead ECG signal with normal beats (N), atrial PB (A) and ventricular PB (V); (b,c,d) represent the constructed by the modified limb leads II and V1 single-plane VCG signal for N, A, V beats, respectively. The measurement of the parameter VCGAng is demonstrated.

Interbeat RR-interval Difference. This parameter is calculated according to equation (1) and represents the difference between the durations of the two RR intervals, surrounding the tested heartbeat (with index n). The normalization towards the mean value of the previous five consecutive RR intervals is applied, in order to achieve a value independent from the heart rate.

(1) $$RRDiff_n = \frac{RR_n - RR_{n-1}}{\left(\sum_{i=n-7}^{n-2} RR_i\right)/5}.100\,(\%)$$

QRS-Width Difference. The QRS-Width is measured as the time-interval between the onset and the offset of the identified QRS pattern. Then the difference between the width of the tested heartbeat (with index n) and a reference width $QRSWidth_{REF}$ (defined below) is calculated according to equation (2).

(2) $$QRSWidthDiff_n = \frac{\left|QRSWidth_n - QRSWidth_{REF}\right|}{QRSWidth_{REF}}.100\,(\%)\cdot$$

QRS-Area Difference. The QRS-Area is calculated between the onset and the offset of the identified QRS pattern as the sum of the absolute values of the ECG samples in the two ECG leads. Then the difference between the area of the tested heartbeat (with index n) and a reference area $QRSArea_{REF}$ (defined below) is calculated according to equation (3).

(3) $$QRSAreaDiff_n = \frac{|QRSArea_n - QRSArea_{REF}|}{QRSArea_{REF}}.100\,(\%)\cdot$$

VCG-Angle Difference. VCG-Angle is the angle of the maximal amplitude of the QRS vector in the single vectorcardiographic (VCG) plane formed by the two ECG leads. Then the difference between the VCG-Angle of the tested heartbeat (with index n) and a reference angle $VCGAng_{REF}$ (defined below) is calculated according to equation (4).

(4) $$\mathbf{V}ECGAngDiff_n = |QRSAng_n - QRSAng_{REF}|\ (\text{Deg})\cdot$$

For the sake of demonstration the *VCG-Angles* of N, A and V-beats are presented in fig. 2(b-d).

Heartbeat classification

In the classification stage, beat-by-beat classification is performed, using a set of rules for the discrimination of the following 3 categories: (i) N - normal or predominant beats; (ii) A - premature atrial beats; (iii) V - premature ventricular beats. A detailed description of the different tests, which are applied in sequence to detect the existence of a defined atrial arrhythmia type (see fig. 1) is given below.

- *Test for Absolute Arrhythmia.* The absolute arrhythmia is usually observable during propagation of atrial flutter/fibrillation when no atrial PBs are present. The algorithm should recognize this state of disorder in the QRS appearance by analysis of the RR-intervals and their differences and should disable the detection of A-beats. The developed algorithm operates with 10 consecutive RR-intervals, calculates their mean value and estimates the normalized difference between each RR-interval and the mean value (labeled below as ($RR\text{-}Mean_{RR}$), in percents). Then we define 10 windows between:

($RR\text{-}Mean_{RR}$) - 1% and ($RR\text{-}Mean_{RR}$) + 1%.

The overlapping windows are merged in a common interval. In case of more than 6 intervals, an absolute arrhythmia state is detected, and only N-beats or V-beats could be present.

- *Test for Supraventricular Tachycardia.* The supraventricular tachycardia has a rapid frequency above 120 bpm, and is generated by a center above the branching of the common bundle and, therefore capable to initiate a beat that could be conducted normally through the ventricles. The proposed algorithm checks for the first RR<0.5s, marks the onset of the tachycardia and calculates the reference values $QRSArea_{REF}$ and $VCGAngle_{REF}$ as the medians estimated for the five preceding QRS complexes. All subsequent heartbeats with shortened RR-intervals (RR<0.5s) are compared with the already computed $QRSArea_{REF}$ and $VCGAng_{REF}$ by substituting them in equations (3) and (4). The derived parameters *QRSAreaDiff* and *QRSAngDiff* are used for discrimination between complexes with atrial and ventricular origin. The QRS complex is labeled as A-beat when:

$$QRSAreaDiff < 50\% \text{ and } QRSAngDiff < 25\text{Deg},$$

otherwise V-beat is detected.

- *Test for Pathologic Pause.* The appearance of two or more RR-intervals with enhanced duration (>1.5s) in 15s ECG segment is a sign for sick sinus syndrome. Such prolonged pause could be interrupted by escape from a lower site cells with pacemaking capability. When the escape pacemaker is located in the atria, the initiated heartbeat is labeled as A-beat. The algorithm is designed to search for two pathologic pauses within 15s and all heartbeats between them are checked whether they are A-beats or V-beats using the same criteria as in the '*Test for Supraventricular Tachycardia*' branch. Here, the reference *QRS-Area* and the reference *VCG-Angle* are determined to be equal to the area and the angle of the first QRS complex after the pathologic pause.

- *Test for Single Premature Heartbeat.* Premature beats (atrial or ventricular) feature with early appearance and prolonged RR-interval until the next normal beat. The premature atrial beats are usually followed by an incomplete compensatory pause. They could be easily recognized by the difference of the coupling intervals, assessed by the parameter in equation (1) *RRDiff>15%*, with the following additional restriction conditions:

$$\begin{cases} RR_{n-1} < 0.9\left(\sum_{i=n-7}^{n-2} RR_i\right)/5 \ \text{or} \\ RR_n > 1.1\left(\sum_{i=n-7}^{n-2} RR_i\right)/5 \text{ and } RR_{n-1} \le \left(\sum_{i=n-7}^{n-2} RR_i\right)/5 \ \text{or} \\ \dfrac{RR_n}{RR_{n-1}} > 1.2 \end{cases},$$

where n is the index of the tested beat.

The abrupt change in the heartrate is also a sign for appearance of premature contraction. Such state is detected with the condition:

$$RR_{n-1} < 0.75\left(\sum_{i=n-7}^{n-2} RR_i\right)/5 \text{ and } RR_n < 0.75\left(\sum_{i=n-7}^{n-2} RR_i\right)/5\,.$$

The detection of premature ventricular heartbeats with the above criteria is not always reliable because in many cases they could be followed by a complete compensatory pause, rarely by an incomplete compensatory pause, or they may appear sandwiched in between two normal beats (the so called interpolated V-beats). This means that when the above conditions for premature beat detection are not satisfied, the output of the algorithm is either N-beat or V-beat.

- *Test for Premature Atrial Heartbeat.* The premature atrial contractions, which have supraventricular origin, resemble the waveforms of the normal or predominant beats, while the premature ventricular contractions, originating from beyond the branching of the common bundle, have abnormally prolonged QRS complex (>0.12s) and bizarre waveforms. Therefore, the separation between A-beats and V-beats is performed by estimation of the morphological parameters *QRS-Width, QRS-Area, VCG-Angle* and their differences with reference values, according to equations (2), (3), (4). The reference values $QRSWidth_{REF}$, $QRSArea_{REF}$, $VCGAng_{REF}$, are calculated as the median value of the respective parameter estimated for the five previous QRS complexes. The defined conditions for A-beat recognition are as follows:

$$\begin{cases} QRSWidthDiff_n < 5\% \text{ or} \\ QRSWidthDiff_n < 10\% \text{ and } QRSAreaDiff_n < 25\% \text{ or} \\ QRSAngDiff_n < 25Deg \text{ and } QRSAreaDiff_n < 50\% \end{cases}$$

If the defined above conditions are not satisfied, the QRS complex is labeled as V-beat.

Results

The described above algorithm for detection of atrial PBs was implemented in software utility, using the software package MATLAB 6.0. The method was developed and tested with 2-lead ECG recordings, including all 48 files from the MIT-BIH arrhythmia database [15], with duration of 30 min per ECG recording. The original database annotations were accepted and the beats were divided into 3 general categories, in accordance with the aim of our study:

- Type N - include all beats, which appear as predominant beats in the ECG recording - originally annotated as: normal beats, left or right bundle branch block and paced beats. The group include a total number of about 97100 cases;
- Type A - annotated as atrial premature beats with a total number of about 2540 cases;
- Type V - annotated as ventricular premature beats with a total number of about 6970 cases.

The classification rules of the algorithm, that involve the parameters *RRDiff*, *QRSWidthDiff*, *QRSAreaDiff* and *VCGAngDiff*, were set on the basis of statistical evaluation of these parameters for all beats in the defined above heartbeat categories. The results for the median value, 10% - 90% range around the median value and the entire range of the defined above RR interval and QRS morphological characteristics are illustrated in fig. 3.

The accuracy of the algorithm was evaluated with the statistical indices Sensitivity (*Se*) and Specificity (*Sp*), calculated as follows:

$$Se = \frac{Correctly\ Detected\ A\ beats}{Total\ Number\ of\ A\ beats}$$

$$Sp = \frac{Correctly\ Detected\ (N\ \&\ V)\ beats}{Total\ Number\ of\ (N\ \&\ V)\ beats}$$

The testing of the algorithm presented a relatively high accuracy with *Se=92.2 %* and *Sp=96 %.*

The performance of the algorithm is illustrated with examples of different ECG arrhythmias shown in fig.4. It is necessary to compare the detection labels (marked in the top of each subplot) with the annotation labels (shown between the two ECG leads). The examples are representative for some specific and complicated cases, which introduce difficulties in the classification of the heartbeats.

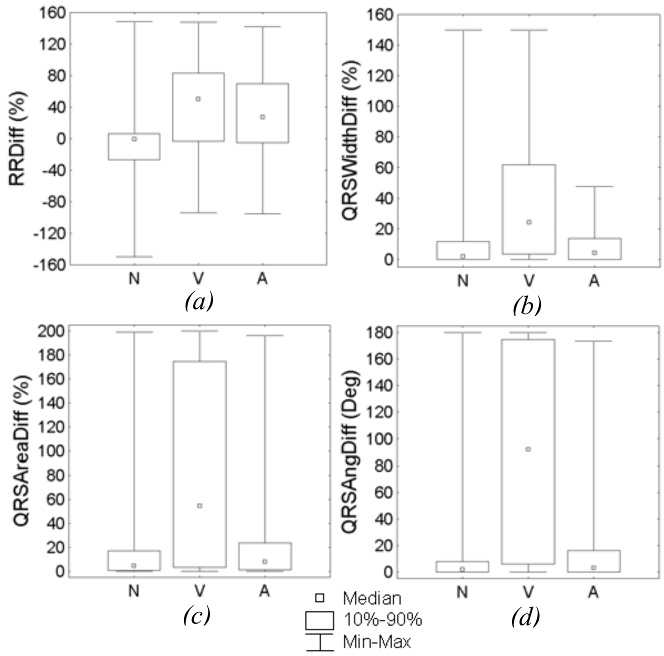

Fig.3. Statistical assessment of RR interval and QRS morphological characteristics for normal (N), premature ventricular (V) and premature atrial (A) beats.

Discussion and Conclusion

The reliability of the algorithm for detection of A-beats is due to the selection of appropriate parameters, which assess the differences between RR intervals and QRS morphological characteristics of adjacent beats. It is evident from fig.3(a) that the N-beats appear normally in regular RR intervals, which results in median value of the *RRDiff* around zero. Every N-beat just before or just after a premature contraction (A or V) characterizes with a negative value of *RRDiff*, due to the usual compensatory pause. These cases result in expansion of the *RRDiff* range (10%-90%) for N-beats towards -30 %. In contrast, the earlier appearance of the premature contractions leads to positive *RRDiff* values (usually up to 80%) both for A-beats and V-beats. The cases with *RRDiff* around zero are most likely to correspond either to the regular escape A-beats or V-beats between two pathological pauses (fig.4d), or to tachycardias - supraventricular (fig.4c) or ventricular, respectively. Therefore, we applied a rule *RRDiff*>15% to recognize the single premature beats, while different rules are set for detection of the other arrhythmias containing A-beats.

All premature beats (A and V) showed overlapping ranges for the *RRDiff* parameter. Therefore, for their separation we relied on the analysis of the QRS morphological descriptors (fig.3 (b-d)). As expected for A-beats, the descriptors, which represent the morphology differences between adjacent QRS complexes, characterize with well-grouped distributions in the range of the low-values (*QRSWidthDiff* less than 10 %, *QRSAreaDiff* less than 25 % and *VCGAngDiff* less than 20 Deg). In contrast, the V-beats feature with wide distributions in the ranges up to 60 % for *QRSWidthDiff*, up to 180 % for *QRSAreaDiff* and up to 180 Deg for *VCGAngDiff*. Taking into account these particularities of the statistical distributions, we defined threshold rules, which effectively served for separation between A and V beats, as it is observable in the example of fig.4(a).

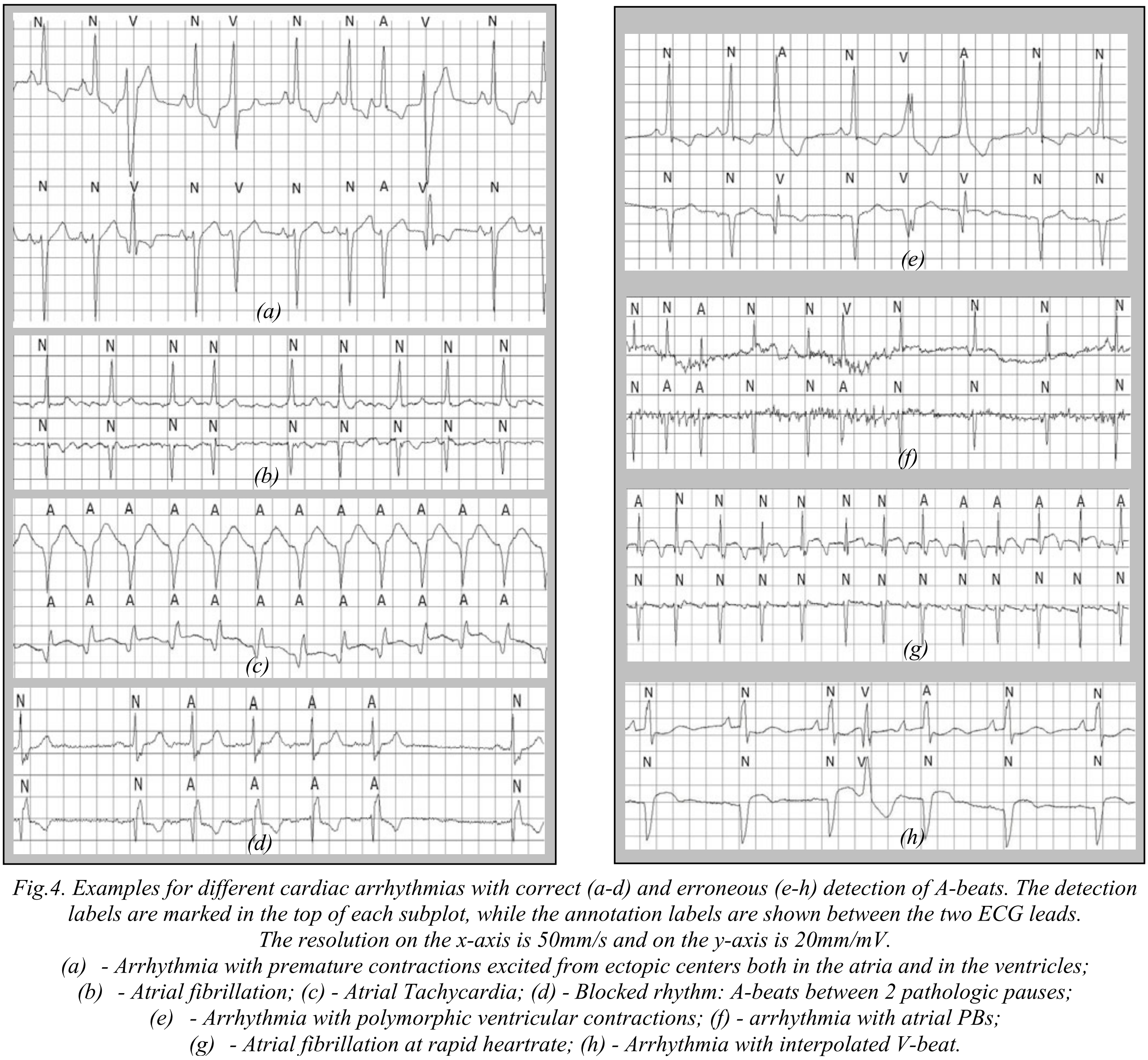

Fig.4. Examples for different cardiac arrhythmias with correct (a-d) and erroneous (e-h) detection of A-beats. The detection labels are marked in the top of each subplot, while the annotation labels are shown between the two ECG leads. The resolution on the x-axis is 50mm/s and on the y-axis is 20mm/mV.
(a) - Arrhythmia with premature contractions excited from ectopic centers both in the atria and in the ventricles;
(b) - Atrial fibrillation; (c) - Atrial Tachycardia; (d) - Blocked rhythm: A-beats between 2 pathologic pauses;
(e) - Arrhythmia with polymorphic ventricular contractions; (f) - arrhythmia with atrial PBs;
(g) - Atrial fibrillation at rapid heartrate; (h) - Arrhythmia with interpolated V-beat.

We proved that the studied parameters become a reliable set for detection of A-beats (see fig.4(a-d) for some examples of typical arrhythmias with true positive detections). However, some exceptions from the classification rules resulting from the wide variety of individual cardiac dysfunctions are observable. For example, some of the polymorphic V-beats could resemble the waveform of the normal contractions, thus leading to false positive detection of A-beats (fig.4e). The false negative classifications are often due to noise in the ECG channel, e.g. the A-beat recognized as V-beat in the corrupted by tremor ECG in fig.4(f). An evident cause for false positive errors of the algorithm is the problematic recognition of absolute arrhythmia in recording with atrial flutter/fibrillation at rapid rate and relatively regular RR intervals, like the example in fig.4(g). In this case, a rush of atrial tachycardia with frequency above 120 bpm is detected. Another reason for false positive errors is the interpolated V-beats with shortened coupling RR intervals, which predisposes to false classification of the next N-beat as A-beat (fig.4h).

The developed algorithm is applicable as quasi real-time operating software, e.g. for fast analysis of 24h holter ECG recordings. It could be implemented as a branch of a more complicated method for heartbeat classification. It is highly effective for detection of A-beats (between 5% and 15% higher sensitivity than other reported algorithms), but an additional approach for discrimination between N and V beats is advisable for extended classifications.

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