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A LINEAR FILTER BANK APPROACH TO DISTINGUISH BETWEEN NORMAL SINUS RHYTHM AND ATRIAL FIBRILLATION

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ABSTRACT

Heart rate variability (HRV) is an indicator of cardiac health. In fact, it provides a powerful tool for observing the relationship between the sympathetic and parasympathetic nervous systems. For HRV analysis RR interval detection is an important step. In our work, eightchannel linear filter bank is designed and its features from analysis filter bank are used for detection of the QRS complexes of ECG signals using five level algorithm. All RR intervals are calculated by using timing information from level five.With these RR intervals, the time domain and frequency domain parameters are calculated. Comparison of all these parameters is done to classify atrial fibrillation from normal sinus rhythm.The study is based on 60 minutes ECG from MIT-BIH Atrial Fibrillation database and MIT-BIH Normal Sinus Rhythm database.

KEYWORDS: ECG, Filter Bank, Heart Rate Variability (HRV), QRS Complex

INTRODUCTION

Heart Rate Variability

The electrocardiogram (ECG) gives cardiac functional details and helps to analyze heartabnormalities. Every heart beat in the ECG signal is a quasi-periodic sequence of P, QRS and T- waves. The QRS complex in this sequence has the highest amplitude and once detected helps in calculating the intervals between consecutive RR peaks.The interval between contiguous QRS complexes is termed as the normal to normal (NN) or the R to R (RR) intervals. Heart rate variability (HRV) refers to the beat-to-beat variations in heart rate. Under resting conditions, the ECG of healthy individuals exhibits periodic variation in R-R intervals.

The HRV measurements are captured noninvasively from the ECG signal. The results from this HRV data are capable of representing physiological condition of the patient and are an important indicator of cardiac disease. Heart rate (HR) is a nonstationary signal and its variation may contain indicators of current disease, or warnings about impending cardiac diseases. The indicators may be present at all times or may occur at random during certain intervals of the time. Atrial fibrillation is typically diagnosed by electrocardiogram (ECG). It is characterized by the replacement of consistent P waves by rapid oscillations or fibrillatory waves, associated with an irregular and often too rapid ventricular response when atrioventricular conduction is intact [1]. Atrial fibrillation is one of the most common cardiac arrhythmias.Above the age of 50, the probability of developing atrial fibrillation doubles with every decade of life. Atrial fibrillation has become a widespread disease in industrialized countries. Atrial fibrillation is not life threatening but the most dreaded complication is embolism, and strokes in particular. The rate of ischemic stroke among patients with atrial fibrillation is two to seven times higher than for people who don’t suffer from atrial fibrillation. One out of six ischemic strokes is caused by atrial fibrillation [2, 3].

- **Analysis of HRV**

Time Domain Analysis

The easiest estimation of HRV is represented by the time domain measures. Simple time–domain variables include the mean RR interval, the mean heart rate, the difference between maximum and minimum heart rate. Recordings for a longer period of 24 hours allow complex statistical time-domain analysis. These statistical parameters may be derived from direct measurements of the RR intervals or from the differences between RR intervals. The simplest variable to calculate is square root of variance i.e. the standard deviation of the NN interval (SDNN). Since variance is mathematically equal to total power under the curve, SDNN covers complete variability in the ECG recording [1].

FREQUENCY DOMAIN ANALYSIS

Similar to the time domain measures of HRV mentioned above, recent developments in microprocessor technology has enabled the calculation of frequency measures on the same ECG data. Frequency measures include the spectral analysis of HRV. The RR interval time series is an irregularly time-sampled signal. This is not an issue in time domain, but in the frequency-domain it has to be taken into account. Therefore, the RR interval signal is usually interpolated before the spectral analysis to recover anregularly sampled signal from the irregularly sampled event series. The HRV spectrum contains the high frequency (0.18 to 0.4 Hz) component, which is due to respiration and the low frequency (0.04 to 0.15 Hz) component that appears due to both the vagus and cardiac sympathetic nerves. Ratio of the low-to-high frequency spectra is used as an index of parasympathetic sympathetic balance [1].

- **Poincare Plot**

A Poincare Plot is generated from all RR intervals. Two consecutive RR intervals represent one point in the plot. The first RR interval (RRN) represents the x-coordinate and the second RR interval (RRN+1) represents the y-coordinate. Poincare Plots for patients with atrial fibrillation look very different. The contours may have shapes like triangles, circles in various sizes, randomly distributed but often concentrated points and any merged contours.

QRS DETECTION METHODOLOGY

Extensive works on the design and use of filter banks is shown in [4,5,6]. Figure 1 shows a FB structure containing a set of analysis filters which decompose the bandwidth of the input signal into sub band signals with uniform frequency bands. The sub bands can be down sampled since the sub band bandwidth is much lower than that of the input signal, processing can be performed on the sub bands according to a specific application. Moreover, the sub bands may be reconstructed by a set of synthesis filters which will perfectly reconstruct the input signal. Figure 1 shows the ideal magnitude responses of the filtersfor m channel filter bank.

The sub bands provide information from various frequency ranges and hence it is possible to perform time and frequency analysis of the input signal. Because the sub bands are down sampled, processing can occur at a lower rate than the input sampling rate. In this work we have used 8 channel filter bank.Initially three features P1,P2 and P3 are computed by using information from sub bands 2,3,4,5 and 6. These features have values which are proportional to the energy of QRS complex [7].

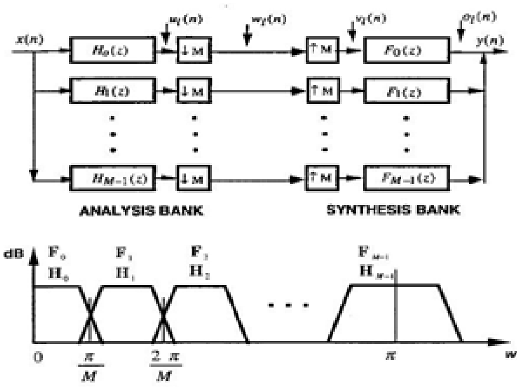

Figure 1: M-Channel Maximally Decimated Filter Bank and Ideal Magnitude Responses of the Filters

Figure 2 shows the sequential levels in the beat detection logic. In this work we have used the five level algorithm for the accurate detection of r peak which was well explained in [7]. The first level detects the beats by detecting peaks in the output of a moving window integration (MWI) on feature P1. Level 2, as shown in Figure 2 has 2 one channel detection processes channel 1 and channel 2 operating simultaneously. Both channels use feature P2 in their respective MWI's, however the preset thresholds are different.Level 3 fuses the beat detection status from each of the 2 one-channel detection algorithms in level 2 by combining a set of if-then-else rules. Level 4 uses another one-channel detection block and uses feature P3 as the input to the MWI. If a beat is detected in level 3, the signal history is updated and the detection status from this level is that the current event is a beat. If level 3 did not classify the current event as a beat, the detection strength of the one channel detection block is computed and compared with the threshold. If the detection strength is greater than the threshold a beat is indicated and the signal history is updated. If the detection strength is less than the threshold the noise history is updated and the detection status from this level is not-a-beat. Level 5 incorporates the timing information necessary for decision logic and to eliminate possible false detection during the refractory period [7].

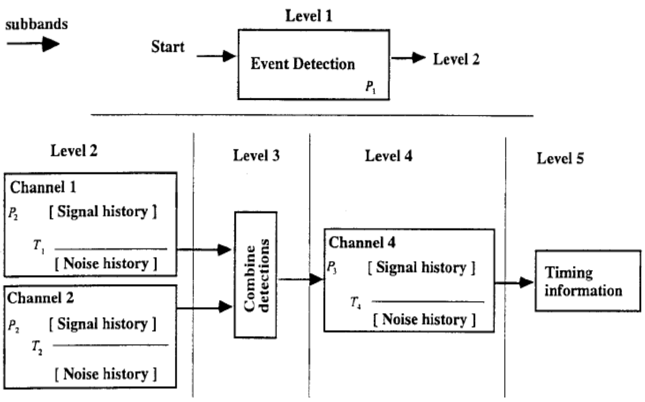

Figure 2: The Overall Beat Detection Logic

RESULTS

For verification, we used ECG signals from the MIT-BIH Atrial Fibrillation Database (MIT-BIH AF DB) and the MIT-BIH Normal Sinus Rhythm Database (MIT-BIH NSR DB), which are derived from the archive of Physio Bank Atm [8]. The MIT-BIH AFDB includes ECG recordings of patients with paroxysmal atrial fibrillation. The MIT-BIH Sinus Rhythm Database consists of ECG records which are originating from healthy persons who do not have significant arrhythmias.

The results of the algorithm are summarized in Tables I. The following resulted from analyzing the ECG recordings from Physio Bank.

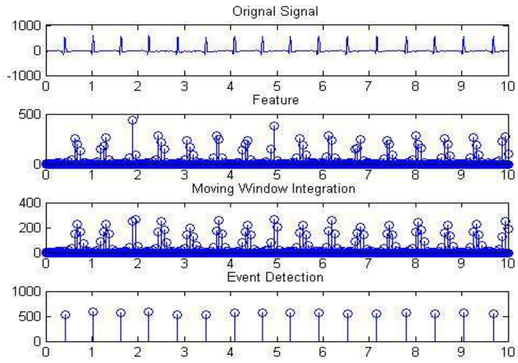

Figure 3: The Plots above Shows the Normal Sinus Rhythm ECG Signal, Computed Feature, MWI Output, Event Detector Output. The Event Detector Flags an Event when a Peak Occurs in the Output of a MWI Operating on Feature P1

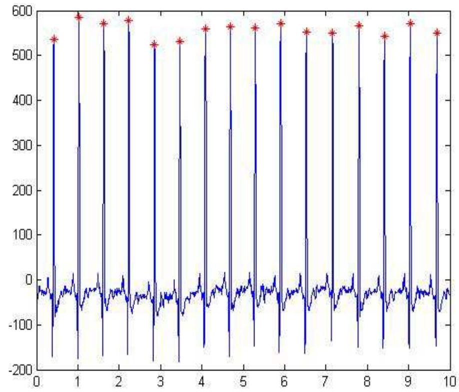

Figure 4: The Plot Shows the Detected r Peaks in NSR Signal for Duration of 10 Sec

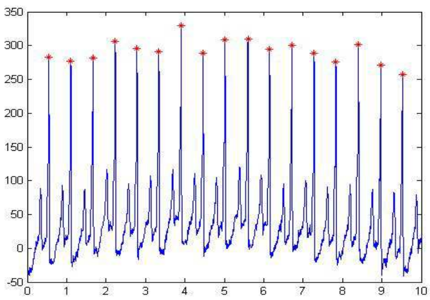

Figure 5: The Plot Shows the Detected r Peaks in AF Signal for Duration of 10 Sec

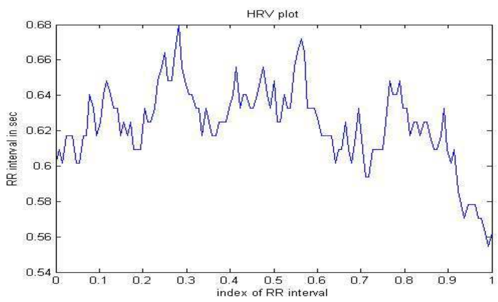

Figure 6: HRV Plot for Normal Sinus Rhythm Signal

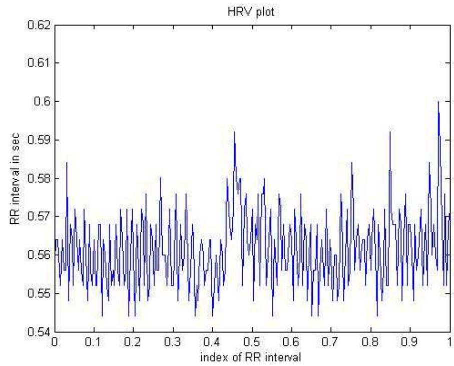

Figure 7: HRV Plot for Atrial Fibrillation Signal

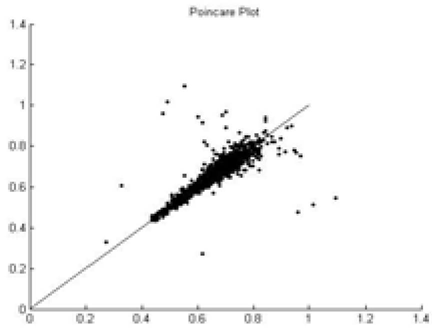

Figure 8: Poincare Plot for NSR Signal

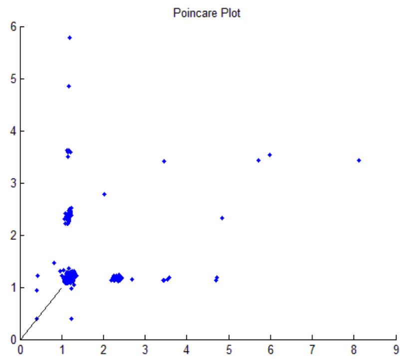

Figure 9: Poincare Plot for AF Signal

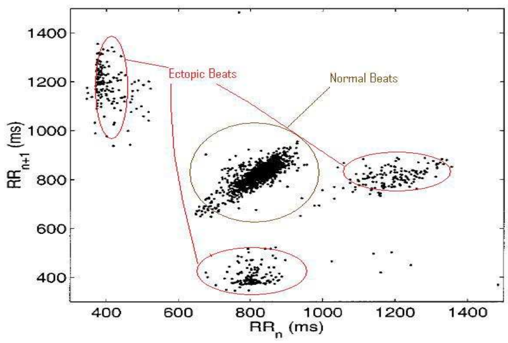

Figure 10: Visualization of Ectopic Beats

Table 1

PARAMETERS\SIGNAL	NSR	AF
SDNN	90.5805±20.8099	217.1073±82.7792
RMSSD	5.5709±2.0398	7.5093±4.7287
pNN50	10.4511±6.1663	37.6699±21.5780
HR	82.3894±10.2626	95.6388±16.4107
LF/HF	1.3851±0.5022	0.7877±0.3055

CONCLUSIONS

The magnitude and duration of QRS waves play an important role in the calculation of HRV. Therefore precise calculation of RR intervals is necessary to depict physiological state. The LF/HF ratio of atrial fibrillation is lower as compared to normal sinus rhythm signal. It indicates that balance between sympathetic and parasympathetic nervous system is good in normal sinus rhythm ECG signal as compared to atrial fibrillation ECG signal. Similarly HR and pNN50 are higher for atrial fibrillation as compared to normal sinus rhythm signal. As in filter bank processing occurs at down sampled rate less computation time is required.

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