Robust Detection of Heart Beats in Multimodal Data Using Integer Multiplier Digital Filters and Morphological Algorithms

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Abstract

We developed a new, robust and efficient heart beat detector in multimodal data using an ECG signal, and one of the pulsatile signals such as blood pressure (BP), if present. To calculate the detection functions, simple and fast integer-multiplier sampling-frequency adjustable digital filters were developed. Using the morphological smoothing, the ECG and pulsatile-signal detection functions, and the noise detection function, are improved. Heart beats are detected using gain-independent adjustable detection thresholds. Streams of detected heart beats in the ECG and in pulsatile signal are linked together. Real time implementation is possible.

1. Introduction

The task of automated detection of QRS complexes of electrocardiogram (ECG) is as old as computing in cardiology. A lot of accurate detectors were developed in the past [1], however there is stil a lack of robust, accurate and noise resistant heart-beat detectors in multimodal data. In the scope of Physionet/Computing in Cardiology Challenge 2014 (Robust Detection of Heart Beats in Multimodal Data) [2] we developed a new, robust, fast, efficient, gain-independent and sampling-frequency adjustable multimodal data heart-beat detector. The detector is capable of operating in ECG signals and in pulsatile signals (P) such as blood pressure, arterial blood pressure, plethysmogram, pulmonary arterial pressure, or photoplethysmogram. The detector was implemented in the C language.

2. Methods

2.1. Detector overview

Initially, the detector seeks for up to three ECG signals and up to three pulsatile, P, signals in a record. During the preprocessing, the detection and noise functions from the ECG and P signals are derived using simple integer-multiplier digital filters [3], and further improved using morphological algorithms [4]. Detecting of heart beats follows in two steps independently in each of the ECG and P signals, and noise intervals are estimated in each of the ECG signals. The detected heart beats in the step I have to pass strict rules. These heart beats are then gathered together from all ECG signals and mapped into a single stream. The ECG signal with the largest number of detected heart beats in the step I is kept for further analysis. Its noise intervals are rejected or shortened accordingly. After that, matching between the streams of detected heart beats in the ECG signal and in each of the P signals estimating the pulse transition times of the P signals, and the matching indexes between the pairs of signals, follows. A P signal with the highest matching index, which also has to satisfy certain matching threshold, is selected for the process of linking the streams of all detected heart beats from the ECG and P signal into one single stream. Finally, the ECG heart beats being detected during the step II, and residing in the noisy intervals, are rejected.

2.2. Feature extraction

We used a class of simple and fast integer-multiplier non-recursive digital filters which can be implemented recursively [3,5]. To extract the QRS complex features, with the aim to derive the ECG detection functions, we designed a slope-sensitive, $H_{\mathrm{E}_1}(z)$, and a peak-sensitive, $H_{\mathrm{E}_2}(z)$, sampling-frequency adjustable band-pass filters:

$$H_{\mathrm{E}_1}(z) = \frac{(1 - z^{\lfloor -5 \cdot F_{\mathrm{S}}/F_{\mathrm{S}_0} \rceil}) \cdot (1 - z^{\lfloor -7 \cdot F_{\mathrm{S}}/F_{\mathrm{S}_0} \rceil})}{(1 - z^{-1})}, \quad (1)$$

$$H_{\mathrm{E}_2}(z) = \frac{(1 - z^{\lfloor -5 \cdot F_{\mathrm{S}}/F_{\mathrm{S}_0} \rceil})^2 \cdot (1 - z^{\lfloor -7 \cdot F_{\mathrm{S}}/F_{\mathrm{S}_0} \rceil})}{(1 - z^{-1})}, \quad (2)$$

where F_{S} is the sampling frequency and $F_{\mathrm{S}_0} = 250$ samples per second. By rounding the exponents in the transfer functions, the filters become sampling-frequency adjustable with pass-bands approx. 10 - 24 Hz. This is the frequency band from which the QRS complexes are mainly composed. The filters are less sensitive to the P and T waves. To derive the noise detection functions, we designed a slope-sensitive, $H_{\mathrm{N}_1}(z)$, and a peak-sensitive band-pass filter, $H_{\mathrm{N}_2}(z)$, with pass-bands approx. 24 - 60 Hz:

$$H_{\mathrm{N}_1}(z) = (1 - z^{\lfloor -3 \cdot F_{\mathrm{S}}/F_{\mathrm{S}_0} \rceil}), \quad (3)$$

$$H_{\mathrm{N}_2}(z) = (1 - z^{\lfloor -3 \cdot F_{\mathrm{S}}/F_{\mathrm{S}_0} \rceil})^2, \quad (4)$$

sensitive to high frequency noises and boundaries of the random shifts. To extract slopes which are the most significant features of the P signals, and to derive the P detection functions, we designed a slope-sensitive band-pass filter, $H_{\mathrm{P}}(z)$, with the pass-band approx. 1.1 - 3.6 Hz:

$$H_{\mathrm{P}}(z) = \frac{(1 - z^{\lfloor -25 \cdot F_{\mathrm{S}}/F_{\mathrm{S}_0} \rceil}) \cdot (1 - z^{\lfloor -50 \cdot F_{\mathrm{S}}/F_{\mathrm{S}_0} \rceil})}{(1 - z^{-1})}. \quad (5)$$

2.3. Detection functions

Figure 1 shows deriving of the ECG detection function, the noise detection function, and detecting noise intervals. To derive initial ECG detection function (Fig. 1.b), $y_{\mathrm{E}}(i)$, where i denotes original signal sample number, the absolute values of the outputs of the filters H_{E_1} and H_{E_2} are added and squared to express the signal energy, and the resulting function is smoothed using a low-pass moving average filter with a cut-off frequency of approx. 6.5 Hz. To further improve the shape of the ECG detection function, morphological smoothing [4] is applied to the $y_{\mathrm{E}}(i)$.

Morphological smoothing of a function f is defined by morphological opening followed by closing using structuring element b [4]. Opening of a function f by a structuring element b is $f \circ b = (f \ominus b) \oplus b$, where $\ominus$ denotes erosion of the f by the element b, $(f \ominus b)(i) = \min_{i \in b}\{f(s+i)\}$, and $\oplus$ denotes dilation of the f by the element b, $(f \oplus b)(i) = \max_{i \in b}\{f(s+i)\}$. Closing of the function f by the structuring element b is $f \bullet b = (f \oplus b) \ominus b$.

To achieve morphological smoothing of $y_{\mathrm{E}}(i)$ structuring element of length of 300 ms was used. The ECG detection function (Fig. 1.d), $d_{\mathrm{E}}(i)$, is than composed following:

$$d_{\mathrm{E}}(i) = \begin{cases} y_{\mathrm{E}}(i) - g_{\mathrm{E}}(i) & \text{if } y_{\mathrm{E}}(i) - g_{\mathrm{E}}(i) \geq 0, \\ 0 & \text{otherwise,} \end{cases}$$

where $g_{\mathrm{E}}(i)$ (Fig. 1.c) is morphologically smoothed version of $y_{\mathrm{E}}(i)$. Morphological smoothing and subtracting reject various fluctuations and bumps at the base of the detection function due to noise and due to T waves, while leaving the shape of the detection pulses unchanged.

The noise function (Fig. 1.e), $n_{\mathrm{E}}(i)$, is derived similarly. The absolute values of the outputs of the noise filters H_{N_1} and H_{N_2} are added and squared, and the resulting output is further smoothed using a low-pass moving average filter with a cut-off frequency of approx. 16 Hz. Morphological smoothing (300 ms) of the noise function, $n_{\mathrm{E}}(i)$, produces the noise detection function (Fig. 1.f), $g_{\mathrm{N}}(i)$, which is a smooth stepwise envelope over the peaks of the noise function in the noisy intervals. Morphological smoothing fills the gaps within these noisy intervals and rejects unwanted spikes left in the QRS complex intervals outside these noisy intervals.

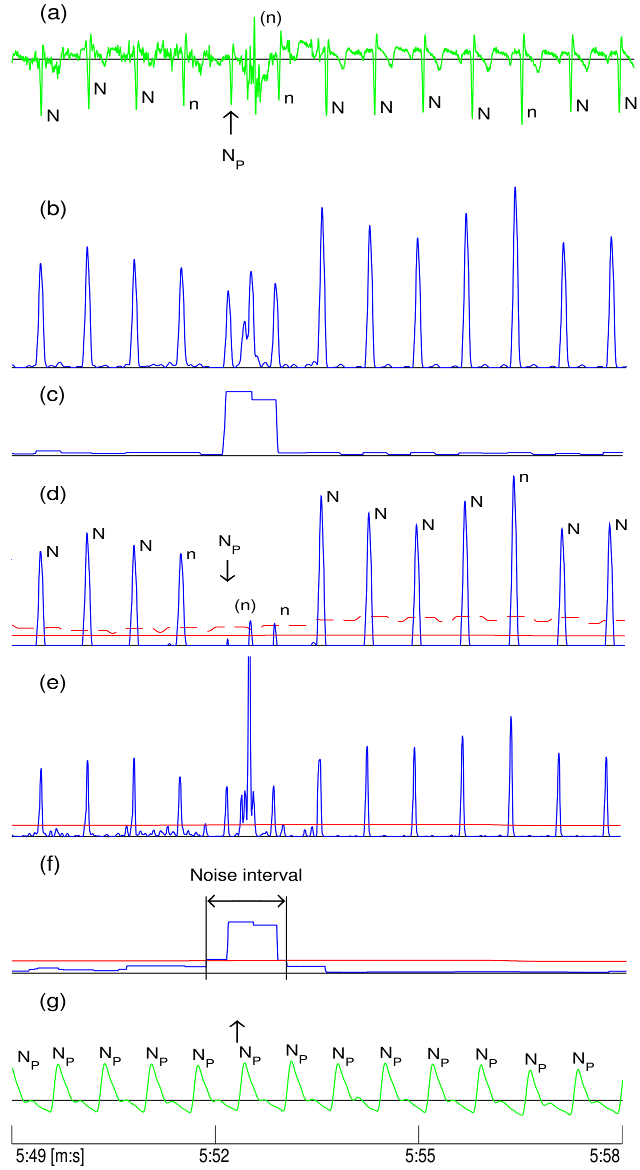

Figure 1. (a) Original ECG signal (excerpt from the record 133 of the training set of the Challenge); (b) Initial ECG detection function, $y_{\mathrm{E}}(i)$; (c) Morphologically smoothed initial ECG detection function, $g_{\mathrm{E}}(i)$; (d) The ECG detection function, $d_{\mathrm{E}}(i)$, with detection thresholds from detection step I, $T_{\mathrm{E}_1}(i)$, (dashed) and II, $T_{\mathrm{E}_2}(i)$, (solid), N_{P} - missed and replaced heart beat, (n) - falsely detected and rejected heart beat; (e) The noise function, $n_{\mathrm{E}}(i)$, with the threshold $T_{\mathrm{E}_2}(i)$; (f) The noise detection function, $g_{\mathrm{N}}(i)$, with the threshold $T_{\mathrm{E}_2}(i)$; (g) A simultaneous P signal (BP) of the record. Upper case N and N_{P}, and lower case n, mark heart beats detected in the steps I, and II, respectively.

Figure 2 shows deriving of the P detection function. The P detection function (Fig. 2.e), $d_{\mathrm{P}}(i)$, is calculated following:

$$d_{\mathrm{P}}(i) = \begin{cases} y_{\mathrm{P}}(i) - g_{\mathrm{P}}(i) & \text{if } y_{\mathrm{P}}(i) - g_{\mathrm{P}}(i) \geq 0, \\ 0 & \text{otherwise,} \end{cases}$$

where $y_{\mathrm{P}}(i)$ (Fig. 2.c) is the output of the filter H_{P} and $g_{\mathrm{P}}(i)$ (Fig. 2.d) is morphologically smoothed (300 ms) version of the $y_{\mathrm{P}}(i)$. Using morphological approach, positive peaks of the $y_{\mathrm{P}}(i)$ are kept while lower positive peaks due to possible isovolumetric relaxation period fluctuations of the P signal are suppressed.

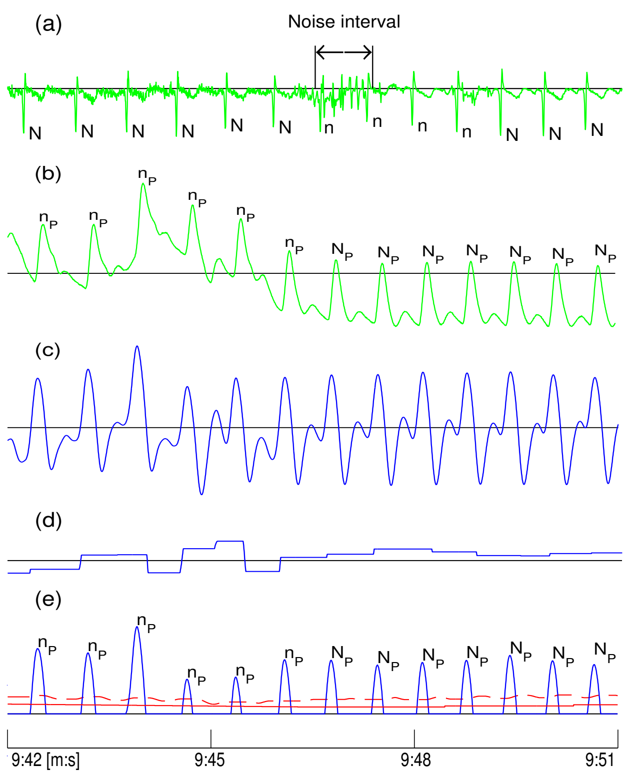

Figure 2. (a) The ECG signal (excerpt from the record 199 of the training set); (b) An original P signal (BP) of the record; (c) Filtered version (by H_{P}) of (b), $y_{\mathrm{P}}(i)$; (d) Morphologically smoothed version of (c), $g_{\mathrm{P}}(i)$; (e) The P detection function, $d_{\mathrm{P}}(i)$, with detection thresholds from detection step I, $T_{\mathrm{P}_1}(i)$, (dashed) and II, $T_{\mathrm{P}_2}(i)$, (solid). Upper case N and $\mathrm{N_P}$, and lower case n and $\mathrm{n_P}$, mark heart beats detected in the steps I, and II, respectively.

2.4. Detection steps I and II

In the detection step I, the detector seeks for as many four similar consecutive pulses in the ECG detection function, $d_{\mathrm{E}}(i)$, as it can find them throughout the record using strict rules. The detection threshold for the step I, $T_{\mathrm{E}_1}(i)$, is set as moving average with a four-second window of the ECG detection function, $d_{\mathrm{E}}(i)$. The differences in the time intervals between the four pulses, and between their peaks, have to be lower than 20%, and 30%, respectively. Besides, the differences in the time intervals between the pulses have to be greater than 0.3 ms (200 bpm) and lower than 1.7 ms (35 bpm). The boundaries are repetitively lowered (5%) for three iterations over entire signal, until success. The four such pulses detected are marked as valid heart beats (N; see Fig. 1 and 2). The time interval of each such four pulses detected is used to calculate the detection threshold for the step II, $T_{\mathrm{E}_2}(i)$. It is calculated as the mean value of the corresponding noise function, $n_{\mathrm{E}}(i)$, within this interval. Besides, the average value of the time intervals between the four pulses defines the ECG heart rate, $\overline{h}_{\mathrm{E}}(j)$, where j denotes the ECG heart-beat number, for these four heart beats. The threshold, $T_{\mathrm{E}_2}(i)$, and the ECG heart rate, $\overline{h}_{\mathrm{E}}(j)$, are linearly interpolated between each next four pulses found. If four consecutive heart beats are not found at least ones, the threshold $T_{\mathrm{E}_2}(i)$ is calculated as four-second moving average of the noise function, $n_{\mathrm{E}}(i)$. The ECG heart beats detected in the step I (N) are judged as final. The noise intervals (see Fig. 1) are detected when the noise detection function, $g_{\mathrm{N}}(i)$, exceeds the threshold $T_{\mathrm{E}_2}(i)$ for the time longer then 0.5 sec.

Similar step-I procedure is applied to the P detection function, $d_{\mathrm{P}}(i)$, producing another stream of detected heart beats ($\mathrm{N_P}$; see Fig. 1 and 2). The only difference is that the detection threshold for the step II, $T_{\mathrm{P}_2}(i)$, is calculated directly from the P detection function, $d_{\mathrm{P}}(i)$. The interpolated detection threshold, $T_{\mathrm{P}_2}(i)$, and the interpolated P pulse rate, $\overline{p}_{\mathrm{P}}(k)$, in pulses per minute [ppm], where k denotes the P signal pulse number, are also calculated. If the detector is not able to find four consecutive pulses at least once over three iterations, such a P signal is ignored.

In the detection step II, every pulse of the ECG or P detection function, $d_{\mathrm{E}}(i)$ or $d_{\mathrm{P}}(i)$, that stays above the ECG or P detection threshold, $T_{\mathrm{E}_2}(i)$ or $T_{\mathrm{P}_2}(i)$, for more than 12 ms, and after that bellow the two thresholds for more than 200 ms, respectively, is marked as valid heart beat, n or $\mathrm{n_P}$, (see Fig. 1 and 2). This step tries to detect heart beats of which detection pulses are by shape less similar to their neighbours.

2.5. Linking streams of detected heart beats

To link the streams of detected ECG and P signal heart beats, matching between the signals must be established in the sense of estimating the pulse shift, $S(k)$, or pulse transition time (PTT), between the j-th QRS complex and the k-th slope feature in the P signal. For this purpose, one of the following two optional methods is used.

The first matching method calculates correlation between the streams of detected heart beats in the ECG and P signal estimating the average pulse shift, $\overline{S}$. The stream of detected heart beats in the P signal is sample-by-sample shifted versus ECG heart beats detected outside the noise intervals, through the correlation window of duration of $\left[0.382 \cdot 60/\overline{h}_{\mathrm{E}}, 0.618 \cdot 60/\overline{h}_{\mathrm{E}}\right]$ seconds. The $\overline{h}_{\mathrm{E}}$ denotes average heart rate of the ECG signal based on heart beats detected in the detection step I. A number of matching heart beats in the ECG and P signal is calculated according to the matching window of length of ± 140 ms, at each sample. The shift with the most matching heart beats represents the average pulse shift, $\overline{S}$.

The second matching method gradually builds a mapping (linear regression function, see Fig. 3) between the interpolated P pulse rate, $\overline{p}_{\mathrm{P}}(k)$, and corresponding pulse shift $S(k)$. For every j-th detected heart beat in the ECG signal, that is outside of the noise intervals, the detector seeks for a matching pulse pair in the P signal through the matching window of duration of $\left[0.382 \cdot 60/\overline{h}_{\mathrm{E}}(j), 0.618 \cdot 60/\overline{h}_{\mathrm{E}}(j)\right]$ seconds. When such a matching pulse pair k in the P signal is found, the $S(k)$ is obtained, and two-column regression table is updated.

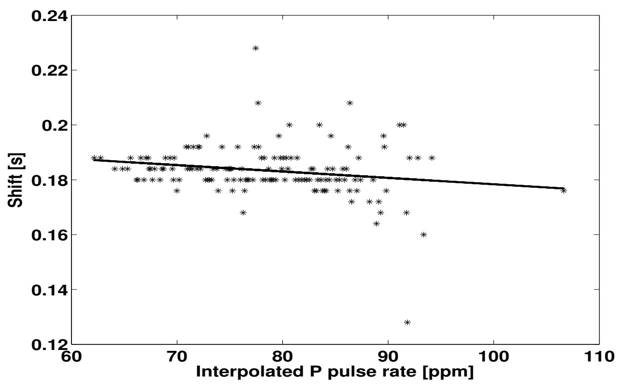

Figure 3. Linear regression function of the record 133 of the training set to map the heart beats detected in the P signal into the stream of ECG heart beats.

During the matchnig procedure, a matching index, M, is maintained as well for each of the P signals. It is defined as the number of matching heart beat pairs in the ECG and P signal versus the number of heart beats detected in the ECG outside the noise intervals. The P signal with the highest M, of which value also have to be higher then 0.75, is selected for the mapping procedure which follows to link the streams of detected ECG and P signal heart beats. If this matching threshold is not met by any of the P signals, only heart beats detected in the ECG signal are kept.

During the mapping procedure, each k-th heart beat detected in the P signal (N_P, n_P) is mapped into the stream of detected ECG heart beats. The mapping is performed only if there is no matching heart beat pair in the ECG signal according to the matching window of length of ± 140 ms. The mapping sets heart beats in the ECG signal that were missed and fills out gaps due to signal loss. An example of mapping of an N_P for a missed heart beat in the ECG signal (within the noise interval) is shown in Fig. 1.d.

Finally, in the noise intervals, the detected ECG heart beats are rejected, unless they were detected in the detection phase I, or they had a matching pair in the P signal during the mapping procedure. An example of a rejected heart beat, (n), which is falsely detected heart beat, is shown in Fig. 1.d.

3. Results

The best scores of the proposed detector over the Phases I, II, and III of the Challenge were 89.24% (2), 85.91% (3), and 85.13% (6). When testing the matching methods (Phase III), i.e., correlation and regression function, the overall scores were almost identical, 83.53% and 83.44%, respectively. Thus, both methods can be considered equivalent. The detector of the entry five of the Phase III was the same as the detector of the entry with the highest score of this Phase (85.13%), but this time, the detector did not seek for the P signals by their label and relied only on the matching index. All but signals labeled as ECG were equally processed as being the P signals. This score was 83.60%.

4. Discussion and Conclusions

The main advantages of the proposed detector are following: reliable feature extraction, improved shapes of the detection functions using morphological algorithms, and the matching method with possible continuous updating of the linear regression function to map between the ECG and P signal heart beats that allows real time implementation.

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