An Adaptive Enhancer with Modified Signal Averaging Scheme to Detect Ventricular Late Potentials

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Abstract. Ventricular late potential detection can be used as a non-invasive diagnostic tool, but traditional detection techniques need around 300 heartbeats and fail to obtain the beat-to-beat information. This paper combines a modified signal averaging and an adaptive enhancer to deal with non-stationary environments and get beat-to-beat information from as little as 60 beats. In the ventricular late potential region of the recovered signal, discernible patterns indicate the presence or not of such waveforms. A maximum absolute value "averaging" can emphasize the boundaries of the QRS complex even further to successfully detect ventricular late potentials.

1 Introduction

Ventricular late potentials (VLPs) are low-amplitude, wideband-frequency waveforms that appear, in the last portion of the QRS complex or beginning of the ST segment, in the high-resolution electrocardiogram (see top plot in Fig. 1) of patients with some life-threatening cardiac diseases. Consequently, the detection of ventricular late potentials can be used as a non-invasive diagnostic marker. However, their detection constitutes a challenge because ventricular late potentials are masked by the other components of the electrocardiogram (ECG) and noise and interference, both in time and frequency domains.

The most commonly used noise reduction strategy for ventricular late potential detection is the coherent averaging [1]. This needs around 300 heartbeats and may introduce a severe low pass filtering effect due to misalignments, not to mention that the beat-to-beat information is completely lost. A modified signal averaging technique [6] [7], which is a combination of mean and median filtering, outperforms the previous way, although the problem of destruction of beat-to-beat information remains.

The standard time-domain analysis employs a high-pass filter to enhance the ventricular late potentials, while attenuating the other components of the ECG and the noise and interference. To avoid ringing and to ensure that the onset and offset of the filtered QRS coincide with those in the original signal, a bi-directional four-pole But-

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terworth high-pass recursive digital filtering is used [4]. This filter, however, cannot be applied in a single direction. In addition, it introduces distortion within the QRS complex. Recently, a finite impulse response filter design based on a parallel combination of all-pass and binomial low-pass filters have been proposed to overcome these problems [6] [7].

A novel adaptive enhancer with modified signal averaging and maximum absolute value "averaging" was designed to obtain certain beat-to-beat information and to emphasize the boundaries of the QRS complex. This facilitates the recognition of patterns in the ventricular late potential (VLP) region, differentiating VLP and non-VLP subjects for diagnosis purposes. The new algorithm can handle certain non-stationary environments and provide beat-to-beat information.

2 Adaptive Enhancer Plus Modified Signal Averaging for Beat-to-Beat Ventricular Late Potential Detection

An alternative time domain analysis strategy for beat-to-beat ventricular late potential detection, based on adaptive line enhancing (ALE) plus modified signal averaging (MSA), was designed here. For ventricular late potential detection, ALE alone may not be good enough [3]. However, combining ALE and MSA, good results have been obtained with less than 64 beats, even for extreme noisy conditions [5]. Here, the initial ALE plus MSA prototype was analyzed and improved.

2.1 Initial Adaptive Line Enhancing Plus Modified Signal Averaging Prototype

Adaptive line enhancing followed by modified signal averaging was proposed for ventricular late potential detection in [5]. Using this approach, it was concluded that the acquisition time can be reduced five-fold to approximately one minute, while maintaining standards in noise reduction for ventricular late potential analysis. However, in a more complete evaluation, increasing the number of real signals, some limitations of this prototype system became apparent.

For some real high-resolution electrocardiographic signals, the system caused troublesome ringing inside the QRS due to instabilities. The concatenation of consecutive windowed heartbeats may introduce discontinuities on the adaptive line enhancing input, due to different levels of the PR and ST segments in the vicinity of the QRS complex. These discontinuities may cause instability in the algorithm, introducing ringing on the output signal. In addition, the same effect may appear associated to the abrupt transitions of the QRS. Consequently, a new adaptive enhancer was devised in an attempt to overcome these limitations.

2.2 New Adaptive Enhancer with Modified Signal Averaging

Fig. 1 shows the new adaptive enhancer with modified signal averaging. The original high-resolution electrocardiographic signal (top plot in Fig. 1) was high-pass filtered (second plot on Fig. 1). This allows a better performance of the adaptive algorithms by diminishing the dynamic range of the input signals, attenuating the drift within the isoelectric segments and enhancing the signal-to-noise ratio in general. In addition, the baseline wandering almost disappears and the filtered segments PR and ST become leveled, avoiding any sharp transition in the further concatenation process.

An enhanced version of the double-level QRS detector algorithm [6] was used to detect N fiducial marks (vertical lines in Fig. 1). Then, the filtered signal was windowed around the marks (100 ms to the left and 156 ms to the right). This can yield a 256N-element vector, by concatenation, and an N-by-256 matrix, x, where every row represents a windowed heartbeat, to perform the modified signal averaging [7].

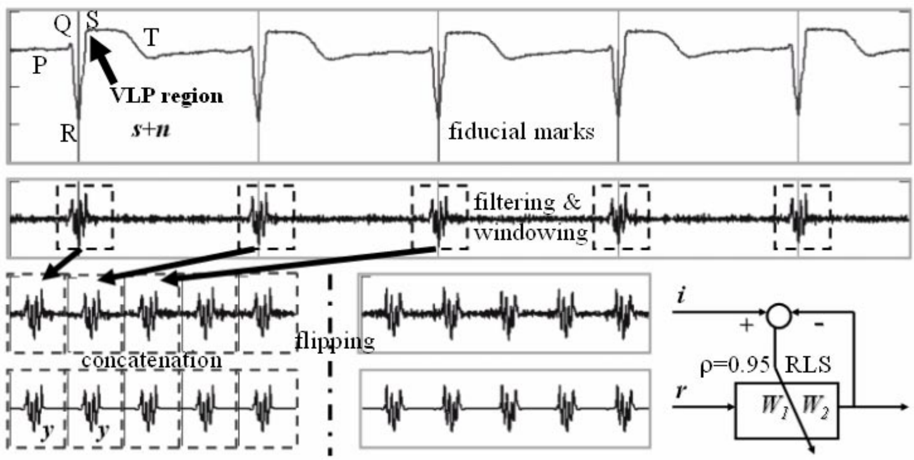

Fig. 1. New adaptive enhancer with modified signal averaging

In the new enhancer, the modified signal averaging [7] was computed to obtain a good quality reference for the adapter enhancer, by repeating N times its output vector y. The main mission of the new enhancer was not to enhance the signal but to track small changes, otherwise lost in the averaging process. This gives certain information on the ventricular late potential beat-to-beat variability that can help in the diagnosis.

Given the matrix x, a matrix m (of size (2d+1)-by-256) is obtained by appending to it d delayed versions of x (each one-column delayed with respect to the previous one) and d advanced versions of x (each one-column advanced with respect to the previous one). Then, the modified signal averaging obtains the output y as

$$y[j] = \frac{\sum_{k=1}^{(2d+1)N} w_{k,j} m_{k,j}}{\sum_{k=1}^{(2d+1)N} w_{k,j}}, \tag{1}$$

where $w_{k,j}$ is the k-th element (row) of the j-th column of the weighting matrix w with similar size as m, and d = 7. The elements of the weighting matrix can be 0 or 1, depending on the median of every column in matrix x (med(x_j)) and the standard deviation of the background noise (σ_{iso}),

$$w_{k,j} = \begin{cases} 1 & if \quad \left| m_{k,j} - med(x_j) \right| \le 2\sigma_{iso} \\ 0 & otherwise \end{cases} . \tag{2}$$

Due to the steep transition between the large QRS complex and low level ventricular late potentials, it was "conjectured" that filtering from the end of the data toward the beginning would give better results. Consequently, a time reversed or "flipped" filtering scheme was adopted.

A flipping operation yields the main input (i) and the reference (r) for the adaptive enhancer. By flipping the concatenated-vectors over, the high-resolution electrocardiogram sequence is processed in the backward direction. This processing from left to right achieved better stability (less ringing and distortion), and good tracking of the ventricular late potential changes.

A second-order RLS adaptive scheme, with a forgetting factor of 0.95, was used to "enhance" the signal i with the reference r. The adaptive enhancer here does not reduce noise in the isoelectric segment compared to the reference r, but it detects certain changes in the ventricular late potential segment. It should be mentioned, to be precise, that the system cannot follow every change in i because of the limitations of the reference r, but it gives a good idea of the variability.

The RLS algorithm allows a fast adaptation to any variation in the signal. By using a forgetting factor ρ of 0.95, the data in the distant past are forgotten [2] and the enhancer can cope with a certain degree of non-stationarity. A filter of order two was found as a good compromise to provide an acceptable frequency separation (long enough), with a quick convergence and low computational load (short enough). It was found that the algorithm converged during the first windowed heartbeat. This enhancer can track not only amplitude variations, but also displacement or phase variations.

The vector at the output of the adaptive system has to be flipped to recover the normal forward direction. This recovered vector can be written as an N-by-256 matrix, o_2, by taking every heartbeat as a different row. Finally, o_2 can be used to obtain an "averaged" vector o_3 by means of a maximum absolute value (MAV) operation. To avoid the influence of outliers, the samples of every column of o_2 were sorted and the 5% on the top and the 5% on the bottom were trimmed out before applying the MAV. The MAV operation selects, from every column of the matrix o_2 after trimming, the sample whose absolute value is maximum to obtain the vector o_3.

3 Adaptive Enhancer Evaluation

The adaptive enhancer implemented here includes two features that have to be tested. This scheme provides a matrix (output o_2) including certain beat-to-beat information and, at the same time, yields an "averaged" vector (output o_3) that can be used for an overall detection of the ventricular late potential, equivalent to the standard method.

To evaluate the algorithms in very realistic scenarios, a high resolution electrocardiographic (HRECG) database was created. This database includes the HRECG signal from 59 post-myocardial infarction patients and 63 healthy volunteers with no evidence of cardiovascular disease. 5-minute records of the bipolar X, Y, and Z leads from each subject were simultaneously collected at a sample frequency of 1 kHz [6]. Furthermore, simulations of the HRECG signal, ventricular late potentials (fixed-VLP and variable-VLP) and noise were designed to evaluate these algorithms in a more controlled environment [6]. More than a thousand combinations were used for testing.

The quality of a particular segment of the HRECG signal can be expressed in terms of several parameters. Some of the most important parameters used here to qualify a recovered sequence are the variance of noise σ_θ^2, which represents the noise power, the bias b_θ, and the signal-to-noise ratio SNR [6].

3.1 Beat-to-Beat Information

Fig. 2 shows an example of how the novel adaptive enhancer plus modified signal averaging recovers the high-resolution electrocardiographic signal from a noisy environment. The 60-heartbeat records shown in the figure were previously high pass filtered and windowed around the fiducial marks (-100ms/+156ms), to confine the 3-D plots. The time axis was reversed to see the ventricular late potential region (around 150 - 200ms).

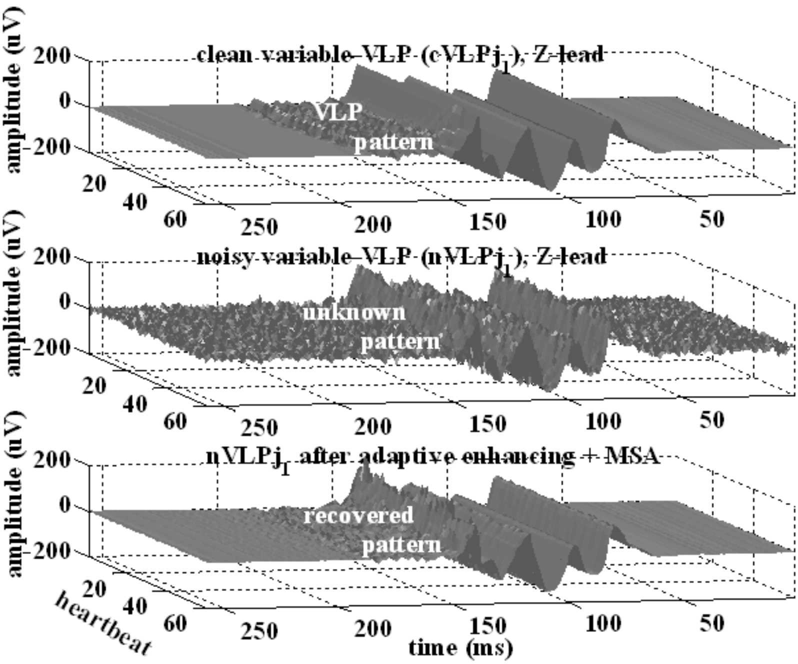

Fig. 2. Example of the performance of the adaptive enhancer with a noisy variable-VLP record

The plot on the top shows a clean record with variable ventricular late potentials (*cVLPj*$_1$). Observe, however, that there is not a recognizable pattern in the region of interest (VLP region, around 150 - 200ms) in the noisy signal $nVLPj_1$ (central plot). The plot on the bottom represents the recovered signal by using the adaptive enhancer plus modified signal averaging (output o_2 explained above). In the recovered signal, there is some distortion close to the steepest regions due to the adaptive algorithm. This distortion avoids following the exact ventricular late potential beat-to-beat structure, which is the main limitation of the adaptive scheme. However, the bottom plot in Fig. 2 clearly shows the pattern of variable ventricular late potentials, which can be distinguished even in a beat-to-beat basis.

As expected, the performance of the algorithm is better for lower levels of noise. It is important to note that, for the same level of noise, the algorithm performs better with non-VLP and fixed-VLP than with the variable-VLP records, although some distortion may be present close to the peaks. Fig. 3 shows an example of that performance with a non-VLP noisy record. Observe the contrast between the enhanced signals (those at the bottom) in the Fig. 2 (variable-VLP) and Fig. 3 (non-VLP).

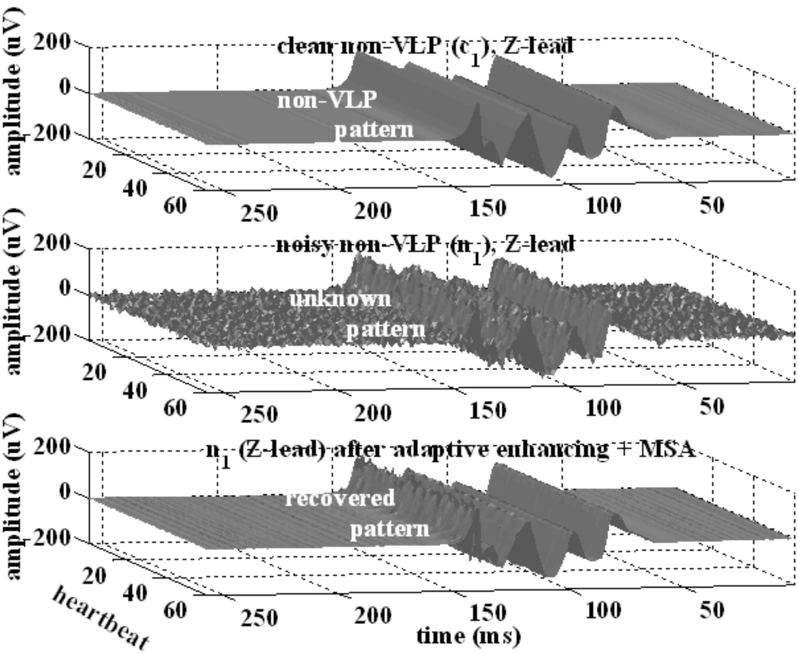

Fig. 3. Example of the performance of the adaptive enhancer with a noisy non-VLP record

3.2 Overall Detection of Ventricular Late Potential

The rationale for the maximum absolute value at the end of the enhancing scheme was that the new modified signal averaging, used to generate the reference i for the adaptive enhancer, worked well in the isoelectric (PR and ST) segments, providing good recovery of that part of the signal, but tended to attenuate the peaks. In the adaptation process, the higher instability is around those peaks because of the worse match between the main input i and the reference input r, and because of the adaptation process itself. The maximum absolute value block does not affect much the isoelectric segment, but it "catches" the peaks (including those in the ventricular late potential region) that are lost otherwise in the averaging process. In this case, the "averaged" vector o_3 does not exhibit better noise reduction in the isoelectric segment than y, but a higher difference between the ventricular late potential and the isoelectric regions, more evident in the presence of variable ventricular late potentials.

Fig. 4 shows the absolute value of the "averaged" 60-heartbeat filtered signal by using the standard method and the adaptive enhancer plus maximum absolute value, i.e. output o_3, compared to the first filtered heartbeat of the ideally clean signal. It can be noticed that the standard averaging method works acceptably well in the isoelectric segments, but attenuates considerably the variable ventricular late potentials (VLP region), making difficult to distinguish the end of the QRS (offset). However, the adaptive enhancing plus maximum absolute value, although introduces some distortion, intensifies the differences between the isoelectric segment and the ventricular late potential region, making easier the recognition of the offset.

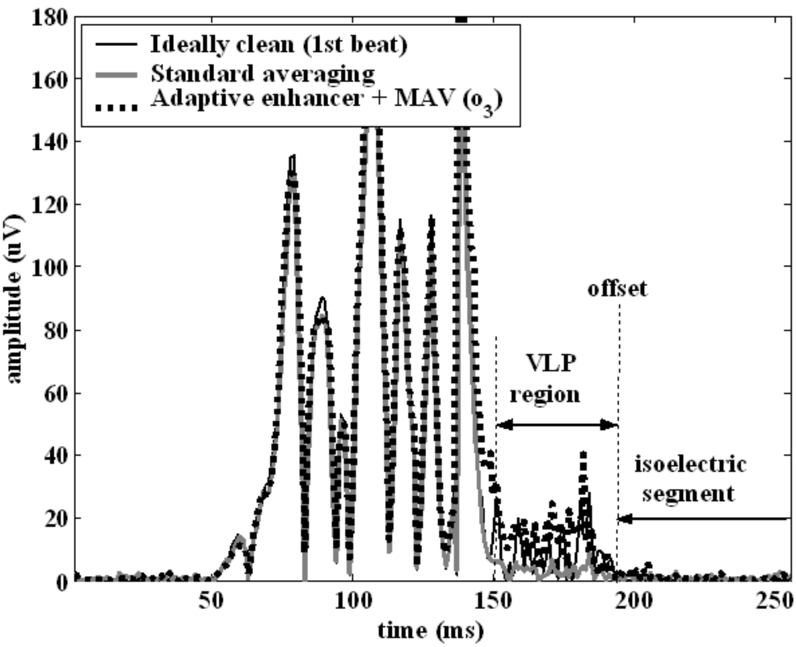

Fig. 4. Absolute value of the "averaged" filtered signals compared to one ideally clean beat

When the ventricular late potentials are fixed (beat-to-beat repeatable), the distortion introduced by the algorithm can be discarded. For these cases, the offset is again easily distinguishable, providing a good discrimination between the VLP and the non-VLP subjects.

A detailed study with 1min test records, showed a perfect classification for the enhancing plus maximum absolute value scheme by using the duration of the QRS as a discriminant feature.

In the previous sections, a limited set of representative figures illustrate the results of the more exhaustive evaluation. However, it was verified qualitative (simple inspection) and quantitatively (computing σ_θ^2, b_θ, and SNR) that the algorithms here presented consistently outperform those reported in [1] and [5].

4 Conclusions

The number of heartbeats needed for the processing algorithms here designed was decreased to less than 60 (i.e. approximately 5 times less than for the standards.) By reducing the acquisition time, the high-resolution electrocardiographic signal is less likely to exhibit non-stationary behavior; nevertheless, the algorithms implemented here have certain capability to handle non-stationary data (forgetting factor < 1 and modified signal averaging which rejects outliers). The new adaptive enhancer plus modified signal averaging provides beat-to-beat information, and different patterns are associated to the VLP region for VLP and non-VLP subjects.

Although some other tests have to be performed before definitively introducing these algorithms to the clinic application, the results so far show a great improvement in the sensitivity and specificity. The processing techniques assessed, outperformed the classical time-domain analysis method. Improved processing algorithms to detect and analyze ventricular late potentials allow for better diagnosis capabilities. Therefore, the results of this work can have a direct impact on the lives of many individuals.

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