An Improved Weighted Signal Averaging Method for High-Resolution ECG Signals

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Abstract

In this paper, we propose a new technique to estimate the coefficients for the weighted averaging of high-resolution ECG (HRECG) records. The purpose of this study was to minimize the number of beats to average in records contaminated with non-stationary noise.

The following averaging methods were studied: 1) Linear averaging of all beats (LA), 2) Linear averaging rejecting very noisy beats (RA) and 3) Weighted averaging (WA). For WA, two techniques for noise variance estimation in each beat were analyzed: a) A noise estimation in a 60 ms window on ST segment (WA-W) and b) A new algorithm that estimates the noise variance from the difference signal obtained by subtracting the averaged beat from the individual beat (WA-D). All methods were tested in simulated HRECG records, contaminated with several types of noise.

We conclude that WA methods minimizes the number of averaged beats compared with LA and RA methods in situations of non-stationary noise. However, the proposed WA-D method shows a higher effectiveness than WA-W method due to a better estimation of the noise variance in individual beats.

1. Introduction

Coherent signal averaging is the classical method to improve the signal-to-noise ratio (SNR) of cardiac micropotentials hidden in the background noise of high-resolution ECG (HRECG) records. It is based on the hypothesis that the signal of interest repeats itself in each beat and that the noise is random and uncorrelated with the signal. The resultant averaged signal is used to detect abnormal cardiac micropotentials, like ventricular late potentials (VLP), which are widely used to identify individuals at high risk of ventricular tachycardia and sudden cardiac death [1].

Electromyographic (EMG) activity is the most important noise source in HRECG records and it has non-stationary characteristics. Therefore, conventional linear averaging is not always as effective as we expect.

In this work, we propose an improved weighted signal averaging method that uses a new algorithm for the noise variance estimation in each beat. The purpose of the method is to minimize the number of beats required to reach a pre-determined noise level in the averaged signal.

2. Methods

2.1. Signal averaging methods

The observed beats in a HRECG record can be modeled as

$$x_k(i) = s_k(i) + n_k(i) \qquad k = 1,\dots,N;\ i = 1,\dots,L$$

where N is the number of beats to be averaged, L is the length of averaging window, $s_k(i)$ is the signal component (deterministic and invariant from beat to beat), and $n_k(i)$ is the zero-mean noise component uncorrelated with $s_k(i)$.

The averaged beat can be expressed as

$$\bar{x}(i) = \sum_{k=1}^{N} w_k x_k(i) \quad \text{with} \quad \sum_{k=1}^{N} w_k = 1$$

Depending on the choice of the coefficients w_k it can be defined three types of signal averaging methods.

a) Linear averaging (LA): all beats are weighted by a factor equal to the inverse of the number of averaged beats.

$$w_k = \frac{1}{N} \qquad k = 1,\dots,N$$

b) Linear averaging rejecting very noisy beats (RA): depending on its noise level, some beats are excluded from the averaging process.

$$w_k = \begin{cases} 1/M & \text{for normal noisy beats} \\ 0 & \text{for very noisy beats} \end{cases}$$

where M is the number of accepted beats and N-M is the number of rejected beats. The criterion used to exclude a grossly noisy beat is based on the signal variance method [2]. In this method, a beat is rejected if its noise level is significantly greater than the mean of the noise levels of all beats previously accepted into the averaging process.

c) Weighted averaging (WA): each beat is multiplied by a coefficient inversely proportional to its noise variance [3].

$$w_k = \frac{1}{\sigma_k^2} \cdot \left(\sum_{j=1}^{N} \frac{1}{\sigma_j^2} \right)^{-1} \qquad k = 1, \ldots, N$$

where σ_k^2 is the noise variance of k-th beat.

2.2. Noise variance estimation

For WA, two techniques for noise variance estimation in each beat are studied:

1) Window estimation (WA-W): the noise variance is estimated in a window located on an isoelectric portion of ECG, like the PR or ST segment, according to this expression

$$\hat{\sigma}_k^{\,2} = \frac{\sum_{j=1}^{W} (x_k(j) - \bar{x}_k)^2}{W} \quad \text{with} \quad \bar{x}_k = \frac{\sum_{j=1}^{W} x_k(j)}{W}$$

where W is the window length and $\bar{x}_k$ is the mean value of k-th beat in the analysis window.

The location and length of the window must be chosen carefully, due to a longer window improves the noise estimation accuracy but can include QRS complex components increasing the noise variance estimation. In this work, we used a window of length W=60 ms, located on ST segment to estimate σ_k^2, previously detrending the data in the analysis window.

2) Difference signal estimation (WA-D): we proposed a new algorithm for noise variance estimation from the difference signal obtained by subtracting the averaged beat from the individual beat.

Assuming that the cardiac signal $s_k(i)$ is deterministic and invariant from beat to beat ($s_k(i)=s(i)$ for $k=1,\ldots,N$), the noise component $n_k(i)$ can be obtained by subtracting the recorded beat $x_k(i)$ from the real cardiac component $s(i)$, that is

$$n_k(i) = x_k(i) - s(i) \qquad i = 1, \ldots, L$$

In practice, the real cardiac signal $s(i)$ is unknown. However, it can be used the averaged beat $\bar{x}(i)$ as an estimator of $s(i)$ to estimate the noise component $n_k(i)$, that is

$$\hat{n}_k(i) = x_k(i) - \bar{x}(i) \qquad i = 1, \ldots, L$$

Thus, the noise variance can be estimated as

$$\hat{\sigma}_k^{\,2} = \frac{\sum_{i=1}^{L} (\hat{n}_k(i) - \bar{n}_k)^2}{L} \quad \text{with} \quad \bar{n}_k = \frac{\sum_{i=1}^{L} \hat{n}_k(i)}{L}$$

where L is the length of averaging window and $\bar{n}_k$ is the mean value of the estimated noise component of k-th beat in the averaging window.

Figure 1 illustrates the proposed method. An individual beat and the averaged beat (obtained previously by linear averaging of 100 cardiac cycles) are represented in Fig. 1a and Fig.1b, respectively. The difference signal shown in Fig.1c is an estimation of noise component of the individual beat.

In order to implement the method, some practical considerations must be taken into account:

I) It is necessary a perfect alignment between the individual beat $x_k(i)$ and the averaged beat $\bar{x}(i)$, to obtain the estimated noise component $\hat{n}_k(i)$. In this study, we have used the cross-correlation method to obtain a synchronization point between both signals.

II) If the cardiac signal determinism condition was strictly met, the window length L could include the total duration of a beat. However, due to small beat-to-beat morphological variations (mainly in the QRS complex), it is advisable do not compute the noise variance in this portion. In this study, we have estimated the noise variance in two windows of 200 ms duration: one of them including the P wave and PR segment and the other including the ST segment and the beginning of T wave.

III) The estimated noise variance $\hat{\sigma}_k^2$ is not exactly equal to the real value σ_k^2, because we have used the averaged beat $\bar{x}(i)$ (which has a residual noise component) instead of the real beat $x(i)$. Due to the residual noise level decreases as the number N of averaged beats increases, it is advisable to use a high number of beats in order to compute the averaged beat. In this work, this beat has been obtained by linear averaging of 100 cardiac cycles.

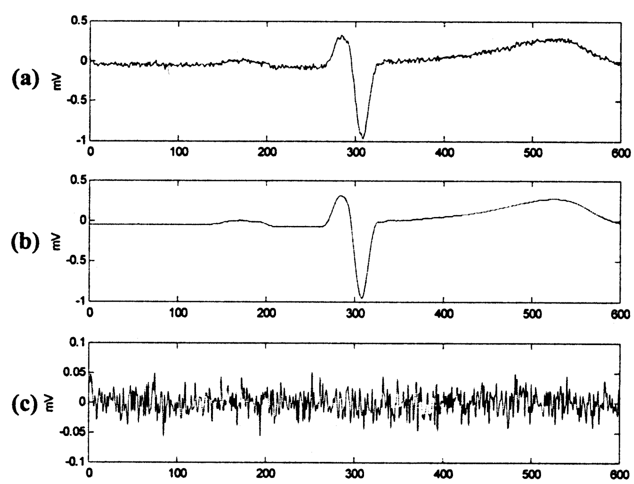

Figure 1. Difference signal estimation. (a) Individual beat, (b) Averaged beat, (c) Difference signal between the individual and the averaged beat.

3. Results

The different averaging methods were tested by applying them to a simulated HRECG record of 300 beats contaminated with several types of additive noise: a) Stationary white noise (SWN), b) Non-stationary white noise (NWN) and c) Real EMG noise, available from the MIT-BIH database. Different RMS levels of the noise records were tested, from 2 to 10μV in 1μV step. The HRECG record was constructed repeating in time a real beat from a HRECG record of a normal subject.

Figure 2 illustrates the initial portion of 60-sec duration of the simulated HRECG record (Fig. 2a) and the different noise records (Fig. 2b-2d).

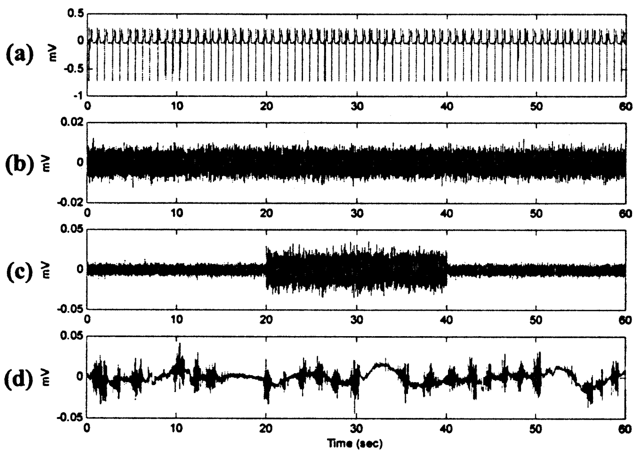

Figure 2. ECG and noise records. (a) Simulated high resolution ECG record, (b) Stationary white noise, (c) Non-stationary white noise, (d) EMG noise.

Both noise variance estimation techniques were tested with SWN and NWN of different RMS noise levels. Figure 3 illustrates the beat-to-beat noise level estimation by both methods for an ensemble of 80 beats contaminated with SWN of 3μV RMS level (Fig. 3a) and NWN with 3μV and 9μV RMS levels (Fig. 3b). It can be appreciated that, in both cases, the difference signal method makes a better estimation of the noise level than the window method.

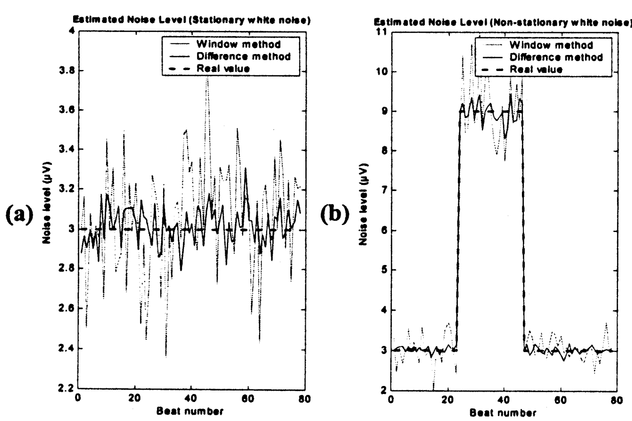

Figure 3. Beat-to beat estimation of the RMS noise level: (a) Stationary white noise, (b) Non-stationary white noise.

The mean ± standard deviation values of the estimated noise levels of 300 beats contamined with SWN of different RMS values are presented in Table 1. In can be observed that the mean value of the estimated noise levels is approximately equal to the real RMS noise level for both methods. However, it can appreciated a lower value of the standard deviation (SD) for the difference method, that implies this method makes a more accurate estimation of the noise level in individual beats.

RMS Noise	Window method		Difference method	
level (μV)	Mean (μV)	SD (μV)	Mean (μV)	SD (μV)
2	2.2762	0.1870	2.0871	0.0911
3	3.1773	0.2952	3.0324	0.1049
4	4.0913	0.3682	4.0387	0.1251
5	4.9793	0.4927	4.9941	0.1557
6	5.9980	0.5849	5.9859	0.1780
7	6.9916	0.6491	6.9722	0.2111
8	8.0274	0.7121	7.9588	0.2307
9	8.9212	0.8245	8.9793	0.2649
10	10.0052	0.9232	9.9841	0.3044

Table 1. Noise level estimation in an ensemble of 300 beats corrupted by SWN of different RMS levels.

In this work, two parameters are proposed to assess the performance of the averaging methods:

a) The number of averaged beats required to reach a 0.3 μV RMS post-averaged noise level. This noise level endpoint maximizes the detection of late potentials in the signal-averaged ECG [4]. This noise reduction must be achieved by averaging from 50 to 300 beats.

b) The final noise level after averaging 300 beats.

The most accurate method of estimating the post-averaged noise level is by the signal variance method [5]. The post-averaged noise level $\bar{n}_R$ is estimated using

$$\bar{n}_R(i)=\sqrt{\sum_{k=1}^{R}\left(x_k(i)-\bar{x}_k\right)^2/R-\left[\sum_{k=1}^{R}\left(x_k(i)-\bar{x}_k\right)/R\right]^2}\Bigg/\sqrt{R}$$

where $x_k(i)$ is the i-th sample value of k-th beat, $\bar{x}_k$ is the mean value of the noise in the measurement window for the beat k (as defined in section 2.2.a), and R is the number of beats included into the averaging process. In this work, the noise level $\bar{n}_R$ is estimated averaging 10 values of $\bar{n}_R(i)$, where the samples i are equally spaced in a measurement window of length W=100 ms on ST segment.

The values of the performance parameters for the different averaging methods for a HRECG record contaminated with SWN, NWN and EMG noise of different RMS noise levels are presented respectively in Tables 2 to 4. The symbol '*' in these tables indicates that the number of beats is higher than 300 beats, which is the maximum number of suggested beats in the standard.

Noise level	Post-averaged noise level (μV)				Beats to 0.3 μV			
	LA	RA	WAW	WAD	LA	RA	WAW	WAD
2	0.1178	0.1178	0.1177	0.1177	47	47	47	47
3	0.1742	0.1742	0.1744	0.1742	98	98	99	98
4	0.2298	0.2298	0.2305	0.2292	172	172	173	171
5	0.2852	0.2852	0.2853	0.2852	269	269	270	269
6	0.3475	0.3475	0.3472	0.3476	*	*	*	*
7	0.4028	0.4028	0.4027	0.4027	*	*	*	*
8	0.4594	0.4594	0.4598	0.4592	*	*	*	*
9	0.5193	0.5193	0.5189	0.5193	*	*	*	*
10	0.5711	0.5711	0.5705	0.5713	*	*	*	*

Table 2. Performance parameters values of all averaging methods for a HRECG record contaminated with stationary white noise of different RMS level.

Burst level	Post-averaged noise level (μV)				Beats to 0.3 μV			
	LA	RA	WAW	WAD	LA	RA	WAW	WAD
4	0.1817	0.1792	0.1764	0.1756	118	117	107	105
5	0.1861	0.1792	0.1770	0.1758	132	117	108	105
6	0.1913	0.1792	0.1771	0.1760	142	117	109	106
7	0.1985	0.1795	0.1775	0.1763	156	118	110	107
8	0.2118	0.1807	0.1780	0.1769	178	119	111	108
9	0.2183	0.1814	0.1786	0.1772	190	121	112	109
10	0.2357	0.1820	0.1798	0.1783	216	122	114	111

Table 3. Performance parameters values of all averaging methods for a HRECG record contaminated with non-stationary white noise of 3μV RMS basal noise level and a burst of 22 beats with different RMS levels.

Noise level	Post-averaged noise level (μV)				Beats to 0.3 μV			
	LA	RA	WAW	WAD	LA	RA	WAW	WAD
2	0.1421	0.1345	0.1322	0.1297	60	55	53	51
3	0.2126	0.2014	0.1982	0.1924	119	109	105	102
4	0.2757	0.2505	0.2461	0.2393	219	190	185	177
5	0.3325	0.3025	0.2960	0.2882	*	*	295	280
6	0.3917	0.3511	0.3422	0.3318	*	*	*	*
7	0.4588	0.4075	0.3935	0.3754	*	*	*	*
8	0.5213	0.4580	0.4413	0.4207	*	*	*	*
9	0.5786	0.5107	0.4921	0.4680	*	*	*	*
10	0.6315	0.5825	0.5587	0.5289	*	*	*	*

Table 4. Performance parameters values of all averaging methods for a HRECG record contaminated with EMG noise of different RMS level.

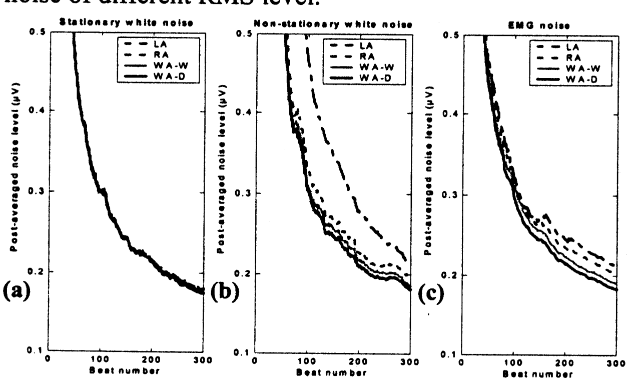

Figure 4. Post-averaged noise level reduction with the number of averaged beats of the averaging methods for: (a) SWN, (b) NWN and c) EMG noise.

Figure 4 illustrates the reduction of the post-averaged noise level with the number of averaged beats of all averaging methods for a HRECG record contamined with: SWN of 3μV RMS level (Fig. 4a), NWN with 3μV RMS basal noise and a burst of 22 beats with 9μV RMS level (Fig. 4b), and 3μV RMS EMG noise (Fig. 4c).

4. Discussion and conclusions

The results indicate that all averaging methods exhibit a similar performance in HRECG records corrupted with stationary white noise. However, the WA method presents a lower value of post-averaged noise level and minimizes the number of averaged beats compared with LA and RA methods in situations of non-stationary white noise and EMG noise. However, the effectiveness of WA method depends on the accuracy of the noise variance estimation in each beat. Usually, the noise variance is estimated in a window located on an isoelectric portion of the ECG record. In this study, we have presented a new method for estimating the noise variance in each beat based on the computation of the variance of the difference signal between the individual beat and the averaged beat. The proposed method allows the use of a longer measurement window than the usual method, making a better estimation of the noise variance and improving the performance of WA method.

We concluded that the presented technique is a promising averaging method for HRECG records. Further investigations should be carried out to examine the performance of this method in real noisy HRECG records.

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