TREMOR SUPPRESSION IN THE ELECTROCARDIOGRAM

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Electrocardiogram recordings are very often contaminated by tremor. Filtering out the tremor is a priori partially successful since it has a relatively wide spectrum, which overlaps the useful ECG frequency band. The proposed method for tremor suppression consists of the following steps. Contaminated ECG signals are entirely subjected to moving averaging (comb filter with linear phase characteristic). The first zero of the filter is set at 50 Hz, thus tremor and power-line interference are together suppressed. The reduced peaks of high and steep waves in the processed signal are then restored by special linearly-angular interpolation using the relations between filtered and non-filtered samples within the QRS complexes. The restored samples are computed by multiplying the filtered samples by a filtered second difference, which is similar to the second difference used for the QRS delineation. The results obtained show satisfactory tremor suppression and low distortions of the QRS peaks.

Keywords: Digital ECG filtering, Tremor suppression, Interference rejection.

1. INTRODUCTION

Electrocardiogram (ECG) recordings are very often contaminated by residual power-line (PL) interference, base-line drift, artifacts, and EMG disturbances due to involuntary muscle contractions of the patient.

Specific digital filter for PL interference cancellation, called subtraction procedure, has been developed some two decades ago and permanently improved later on [1]. It does not affect the signal frequency components around the rated PL frequency. Moving averaging is applied on linear segments of the signal to remove the interference components. They are stored as phase corrections and further subtracted from the signal wherever non-linear segments, e.g. QRS complexes are encountered.

Filtering out the tremor is a priori partially successful since it has a relatively wide spectrum, which overlaps the useful ECG frequency band. One of the first recommendations for ECG instruments [2] suggests a low-pass filter with minimum 35 *Hz* cut-off. However, the amplitudes of sharp QRS waves are reduced. The moving averaging (comb filter with pure linear phase characteristic [3]) gives similar results.

When the comb filter is used as a step of the subtraction procedure [1], the signal inside the QRS complexes is not subjected to moving averaging. Thus, the QRS peaks are preserved but in presence of tremor the complexes become corrupted and the linear segments are not correctly detected, the last leading to: i) unsuppressed disturbance in false non-linear segments, and ii) rare re-calculation of the phase corrections, which can not follow changes of the interference amplitudes. These problems are overcome to some extent by Dotsinsky and Christov [4], who introduced a parallel buffer. The comb filtering is there applied over the entire signal that allow precise locating the linear segments. However, the possibility of denoising the QRS complexes by inappropriate tremor components as a part of the calculated phase corrections still remains.

Another technique for applying the subtraction procedure in case of tremor is reported by Christov [5]. The approach introduces adaptive criterion for linearity detection based on the ratio R between the linear segments length in a selected epoch and its total length usually chosen about 1 s. Normally, the criterion threshold M is a constant, which is set from 100 to 160 μV [1]. In the referred publication [5], M starts from a low value of 50 μV and increases until R reaches a pre-selected value, e.g.0.9 that corresponds to QRS complex and free of noise RR interval with normal dimensions. The results obtained show a reasonable compromise between tremor suppression and QRS amplitudes reduction.

Adaptive filtration has been also attempted but with limited success because the QRS complexes disturb the adaptation process up to the end of the T-waves [6]. Luo and Tompkins obtained faster convergence using additional EMG channel as reference input [7]. Bensadoun et al. proposed a multidimensional method [8] but the reduction of sharp Q-waves amplitudes is too high.

Christov and Daskalov [9] applied a smoothing procedure adopted by Savitzky and Golay [10]. It used least square approximation and introduced 'wings' function for defining the weighting coefficients. The obtained EMG artifact suppression ration is about 6. Reduction of R and S waves from 60 μV to 120 μV is reported depending of the wave shape.

Nikolaev and Gotchev denoised ECG signals by applying wavelet domain Wiener filtering [11]. They mixed original signals with EMG noise with a SNR = 14 *dB*. Two-stage algorithm improves the traditional technique by involving time-frequency dependent threshold for calculating the first stage pilot estimate. A SNR over 20 dB is obtained together with less than 10% QRS amplitudes reduction. In another paper Nikolaev et al. [12] reported an improvement in the SNR of more than 10 *dB*.

2. METHOD AND MATERIAL

The proposed for tremor suppression method consists of the following steps. Contaminated ECG signals are entirely subjected to moving averaging. The first zero of the filter is set at 50 *Hz*, thus tremor and PL interference are together suppressed. The reduced peaks of the processed signal are then restored by special linearly-angular interpolation.

Recordings are taken from the AHA database. They are preliminary moving averaged to suppress any undefined inherent noise. In a first part of the study the obtained conditionally 'clean' input signals are used for development of the interpolation and evaluation of its correctness. For this purpose 'clean' signals are once more comb filtered and then restored. Input and output signals are compared to assess the distortions introduced by the interpolation. In a second part of the study the 'clean' signals are mixed with synthesized PL interference and tremor obtained by two ECG electrodes placed on one forearm. The mixed signals are subjected to the same procedure. The obtained results show the effect of the tremor suppression and PL interference cancellation. In a third part of the study the method is applied on some contaminated by tremor AHA recordings.

3. BACKGROUND OF THE SIGNAL RESTORATION

3.1 Basic relations between filtered and non-filtered samples

The formulae for calculating the middle term during moving averaging over n samples for odd n=2m+1 and even n=2m [1] are presented below:

$$Y_i = \frac{1}{n}\sum_{j=-m}^{m} X_{i+j}, \quad n = 2m+1; \quad Y_i = \frac{1}{n}\left[\sum_{j=-(m-1)}^{m-1} X_{i+j} + \frac{X_{i+m}+X_{i-m}}{2}\right], \quad n = 2m. \quad (1)$$

Taking in consideration that $\sum\limits_{j=-m}^{m} X_{i+j} = \sum\limits_{j=-m}^{-1} X_{i+j} + X_i + \sum\limits_{j=1}^{m} X_{i+j}$ and $\sum\limits_{j=-m}^{-1} X_{i+j} = \sum\limits_{j=1}^{m} X_{i-j}$, eqn. (1) can be expressed by

$$\begin{aligned} Y_i &= \frac{1}{n}\left[\sum_{j=1}^{m}\left(X_{i-j}+X_{i+j}\right)+X_i\right], && n = 2m+1 \\ Y_i &= \frac{1}{n}\left[\sum_{j=1}^{m-1}\left(X_{i-j}+X_{i+j}\right)+\frac{X_{i-m}+X_{i+m}}{2}+X_i\right], && n = 2m \end{aligned} \quad . \quad (2)$$

Substituting $X_i = nX_i - 2\sum\limits_{j=1}^{m} X_i, \quad n = 2m+1; \quad X_i = nX_i - 2\left(\sum\limits_{j=1}^{m-1} X_i + \dfrac{X_i}{2}\right), \quad n = 2m$,

eqn. (2) is transformed in

$$\begin{aligned} Y_i &= X_i + \frac{1}{n}\left[\sum_{j=1}^{m}\left(X_{i+j}-2X_i+X_{i-j}\right)\right], && n = 2m+1 \\ Y_i &= X_i + \frac{1}{n}\left[\sum_{j=1}^{m-1}\left(X_{i+j}-2X_i+X_{i-j}\right)+\frac{X_{i+m}-2X_i+X_{i-m}}{2}\right], && n = 2m \end{aligned} \quad . \quad (3)$$

The polynomial inside the parentheses is a second difference, represents one of the possible versions of the linear criterion [1] and is further denoted as

$D_{i,j} = X_{i+j} - 2X_i + X_{i-j} = \left(X_{i+j}-X_i\right)-\left(X_i-X_{i-j}\right)$.

Using this symbol, the samples X_i and Y_i can be expressed by

$$Y_i = X_i + \frac{1}{n}\sum_{j=1}^{m} D_{i,j}, \quad n = 2m+1; \quad Y_i = X_i + \frac{1}{n}\left(\sum_{j=1}^{m-1} D_{i,j} + \frac{D_{i,m}}{2}\right), \quad n = 2m. \quad (4)$$

The mean signal velocities on the left and the right hand of the ongoing sample X_i are $v_{i,i-j}=\dfrac{X_i-X_{i-j}}{j}$ and $v_{i+j,i}=\dfrac{X_{i+j}-X_i}{j}$. They are averaged within the intervals [*i-j, i*] и [*i*, *i+j*] being related to the time-coordinates *i+j*/2 и *i-j*/2.

Then, eqn. (4) is presented as

$$\begin{array}{ll} Y_i = X_i + \dfrac{1}{n}\sum\limits_{j=1}^{m} j\left(v_{i+j,i}-v_{i,i-j}\right), & n=2m+1 \\ Y_i = X_i + \dfrac{1}{n}\left[\sum\limits_{j=1}^{m-1} j\left(v_{i+j,i}-v_{i,i-j}\right)+\dfrac{m\left(v_{i+m,i}-v_{i,i-m}\right)}{2}\right], & n=2m \end{array} . \quad (5)$$

3.2 Linearly-angular interpolation for back filtering

Let us assume that the original 'free' of noise signal is linear aside from the ongoing sample X_i (Fig. 1) and has a triangular-like shape. Then $v_{i+j,i}=v_r$, $j=i+1,\ldots,i+n$; $v_{i+j,i}=v_l$, $j=i-n,\ldots,i-1$ and the difference $v_{i+j,i}-v_{i,i-j}=v_r-v_l$ as well as the ratio $D_{i,j}/j=D_{i,k}/k$, $k=1,2,\ldots,n$ are constant.

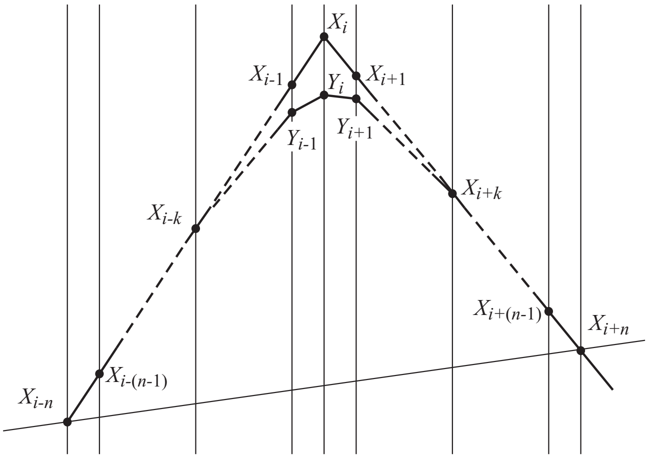


Fig.1. Linearly-angular interpolation of the signal in the interval [i-*n*,..., i+*n*].

The eqn. (5) is transformed to a uniform expression both for odd and even number of averaged samples:

$$Y_i = X_i + (v_r - v_l)\kappa_n = X_i + \frac{D_{i,k}}{k}(v_r - v_l)\kappa_n, \quad (6)$$

where the constant κ_n is given by

$$\kappa_n = \frac{1}{n}\sum_{j=1}^{m} j = \frac{n^2-1}{8n}, \quad n=2m+1; \quad \kappa_n = \frac{1}{n}\left(\sum_{j=1}^{m-1} j + m\right) = \frac{n}{8}, \quad n=2m. \quad (7)$$

Eqn. (6) can be written as

$$X_i = Y_i - D_{i,k}\frac{\kappa_n}{k} = Y_i - D_{i,k}K, \quad (8)$$

Analogously to the second difference $D_{i,k}$, a filtered second difference $D^*_{i,k}$, is introduced using filtered signal samples:

$$D^*_{i,k} = Y_{i+k} - 2Y_i + Y_{i-k}\,.$$

Substituting $D_{i,k} = \eta D^*_{i,k}$, the back filtered sample X^*_i can be calculated by

$$X^*_i = Y_i - \eta D^*_{i,k} K\,, \qquad (9)$$

The coefficient η is intended to consider the shape variety of real signal triangles before the moving averaging. For the time being, this study presumes that η is very close to 1.

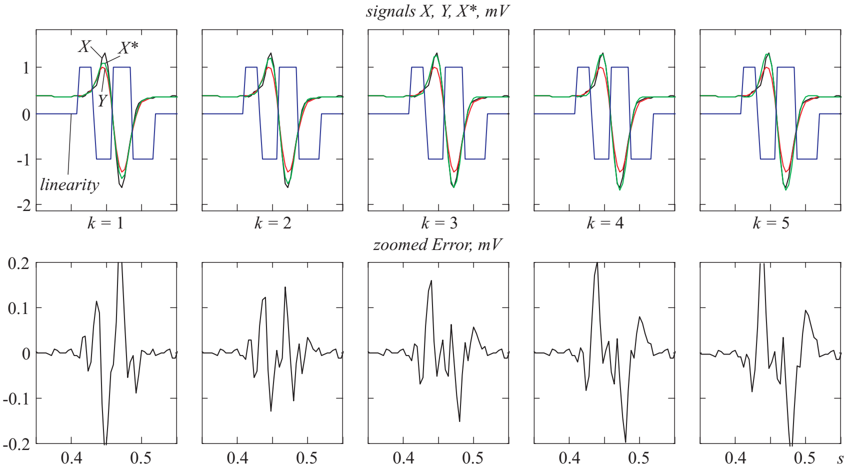


Fig.2. Signal recovering using linearly-angular interpolation.

The influence of k on the back filtering error is assessed by experiments with k=1, 2, 3, 4, 5 and $n = 5$ (Fig. 2) as follows:

1. Conditionally 'clean' signal X is moving averaged.

2. The filtered signal Y is processed by the linear criterion $\left|D^*_{i,n}\right| < M,\quad D^*_{i,n} = Y_{i+n} - 2Y_i + Y_{i-n},\quad M = 0.12\,mV$.

3. The restored signal X^* is obtained by back filtering (eqn. 9), applied on the detected non-linear segments (see the rectangular rebounds in Fig. 2).

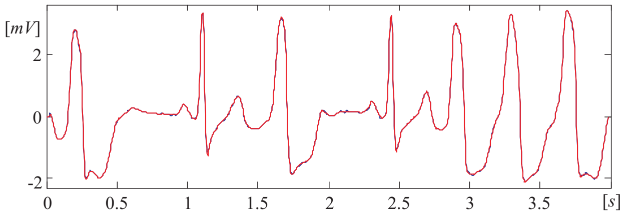
4. The signal distortions are calculated by $err = X^* - X$.

Evidently, the error is minimum with $k = 2$. It can be seen also that the lower values of k contribute to better shape restoration in the neighborhood of the rounded peak while the steeper one is sub-compensated. The higher k values restore well the steep peak but the other one is over-compensated.

4. RESULTS

Four second episodes of some AHA signals are shown in the next Figures. The 'clean' and the restored signals are overlapping in the first subplot. The lower traces demonstrate the error committed.

Actually, the linear segments outside the ventricular beats and any other ECG waves (see for example Fig.3 and 4) that represent physiological zero-line should be free of any distortions. Obviously, the observed 'error' there is due to noise components of the AHA recordings that have been not totally eliminated by the preliminary moving averaging since the first pass-band of this comb filter [3] has the equivalent cut-off of high-pass filter approximately at 25 *Hz*.



Blue trace -> conditionally clean AHA_7009d1 signal; red trace -> restored signal

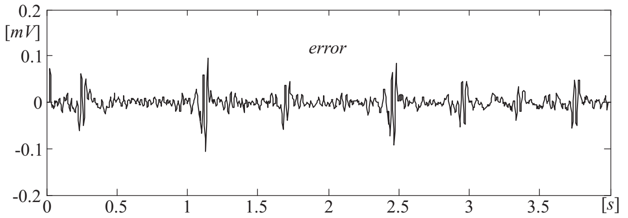
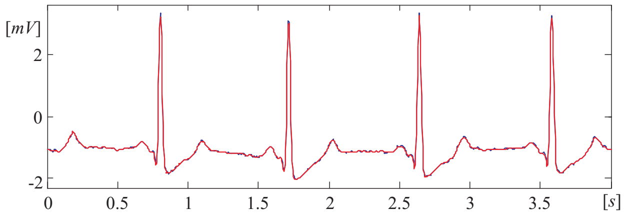


Fig.3.



Blue trace -> conditionally clean AHA_5001d1 signal; red trace -> restored signal

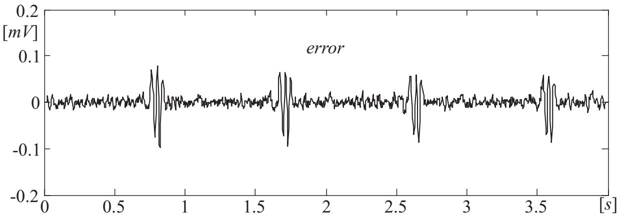
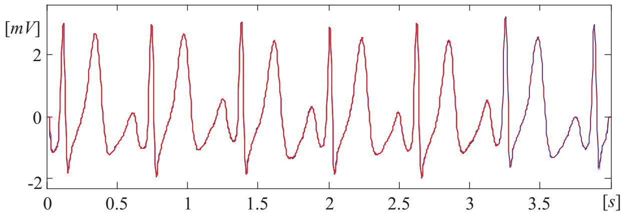


Fig.4.



Blue trace -> conditionally clean AHA_2008d1 signal; red trace -> restored signal

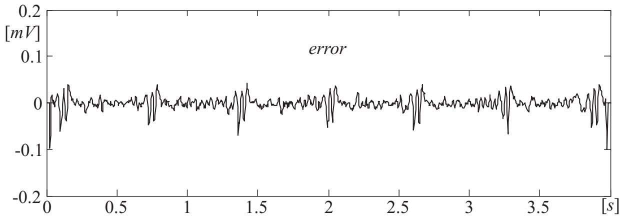
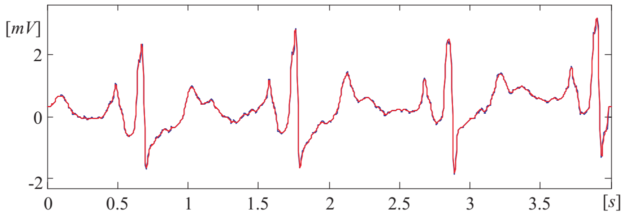


Fig.5.



Blue trace -> conditionally clean AHA_7010d1 signal; red trace -> restored signal

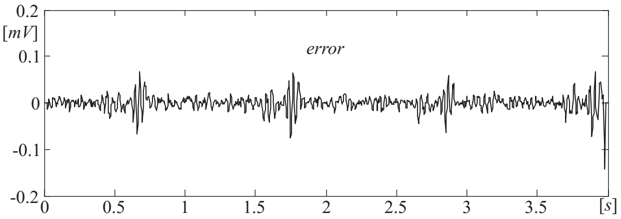


Fig.6.

This is proved indirectly in Fig.7 where an episode of the used tremor before and after moving averaging is presented together with the corresponding equalized in percentage FFT diagrams.

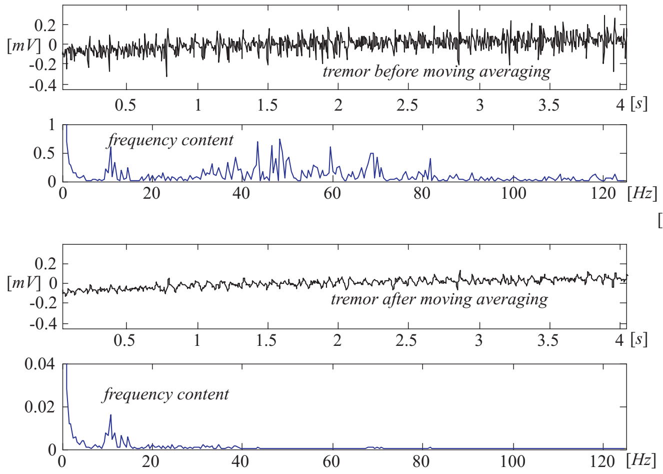


Fig.7.

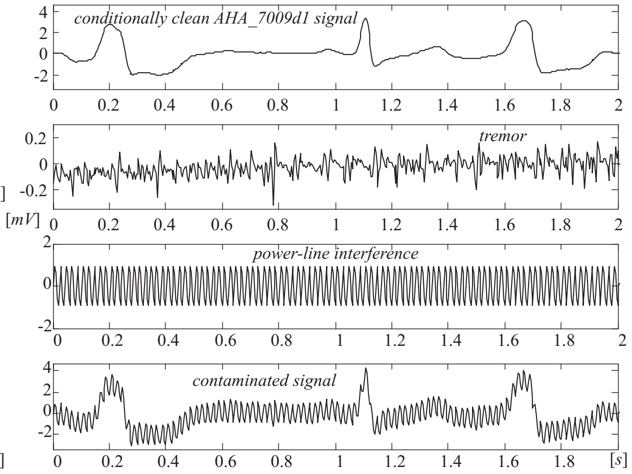
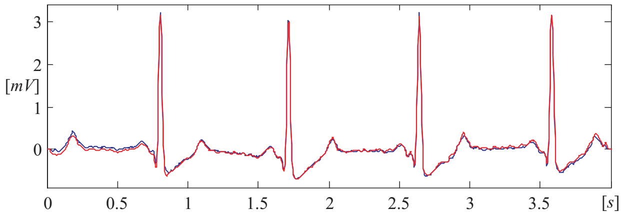


Fig.8.

Consequently, the real errors are considerably smaller. One may speculate that the distortions introduced by the linearly-angular interpolation inside the QRS complexes are within ± 50 μV (see Fig.3-6). The observation of the traces in Fig.7 suggests how to assess the suppression level of the comb filter. It is quite possible that the maximum peak coupled to a relatively high frequency before averaging is well suppressed after averaging while a lower amplitude lower frequency peak before may practically preserve its amplitude after that. Therefore, the suppression level could be defined as the ratio between the maximum peaks in signals before and after averaging. In the episode of question such ratio is over 4 times.



Blue trace -> contaminated AHA_5001d1 signal; red trace -> restored signal

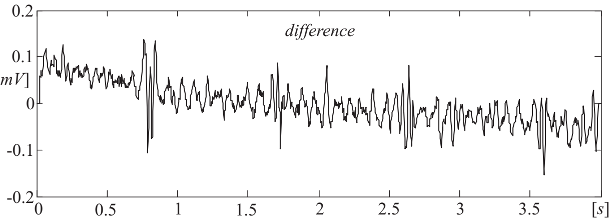
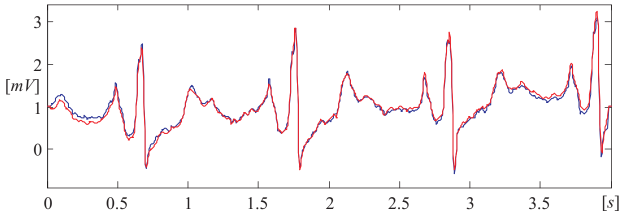


Fig.9.



Blue trace -> contaminated AHA_7010d1 signal; red trace -> restored signal

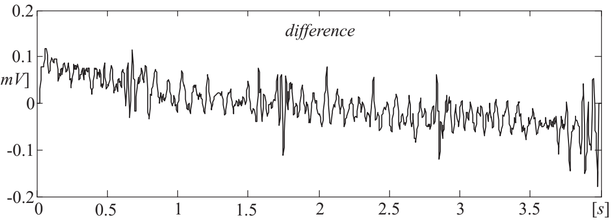


Fig.10.

The mixed (contaminated) AHA_7009d1 signal can be seen in Fig.8 (lower trace) together with the tremor and interference components (second and third traces). Noise suppression of other selected as typical AHA recordings as well as sums (called differences) of extracted tremor and error of restoring the QRS complexes are presented in Fig. 9 and 10.

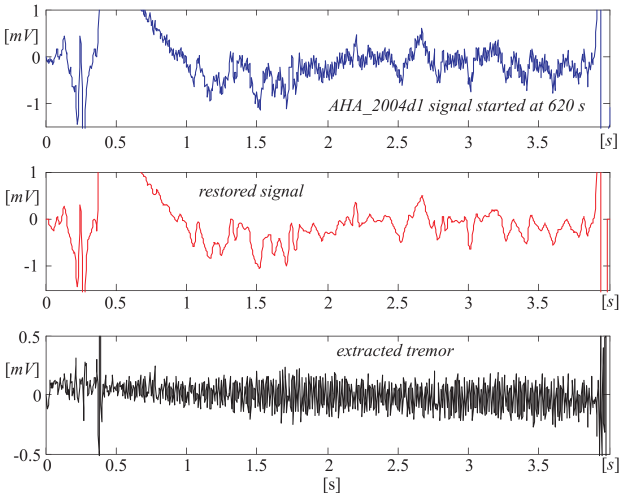


Fig.11.

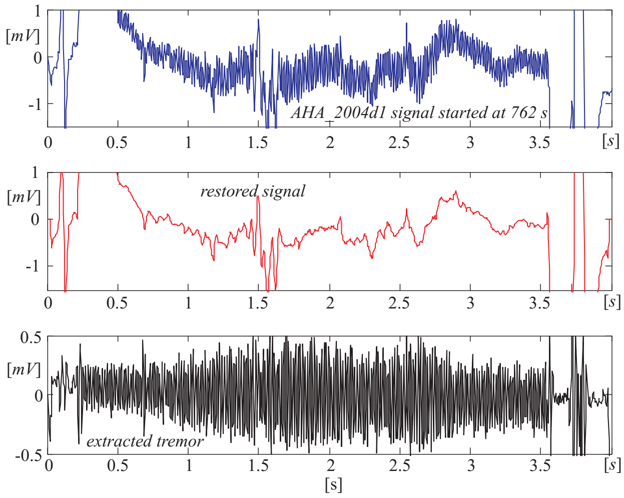


Fig.12.

The PL interference is not discussed because it is totally eliminated [1]. The next Fig. 11 and 12 show the result of applying the procedure on AHA_2004d1 episodes.

The two first traces are original and processed signals, respectively. The lower traces point out the extracted tremor plus small distortions in the QRS complexes due to the linearly-angular interpolation.

4. CONCLUSIONS

The proposed method for tremor suppression is based on total moving averaging of the ECG signal followed by special linearly-angular interpolation for restoring the affected amplitudes of QRS complexes and other relatively high and steep ECG waves. The restoration error is expertly assessed as below the level, which may provoke wrong diagnostic. Therefore, the tremor suppression depends on the type of filter used. The results prove the efficiency of the developed method. For the time being the individual shape of the restored waves (the coefficient η) is not taken in consideration. This possibility will be further checked up. It is suitable also to implement another filter with linear phase characteristic and enhanced level of suppression.

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