SUBTRACTION METHOD FOR REMOVING POWERLINE INTERFERENCE FROM ECG IN CASE OF FREQUENCY DEVIATION

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Extension of the subtraction method for power-line interference removing from ECG signals is proposed in case of power-line frequency variations. The existing version suitable for non-multiplicity between sampling rate and interference frequency is improved and adapted to dynamic deviations around the rated power-line frequency. The ongoing interference values, which are stored in temporary buffer, are used for recalculation of the filter's coefficients. A simplified algorithm is elaborated and experimented in case of odd sample number in one period of the interference. The corresponding formulae for even sample number are also derived. The proposed procedure successfully compensates abrupt and gradual changes of the power-line frequency allowed by the standards.

Keywords: Digital filtering, ECG filtering, Interference rejection.

1. INTRODUCTION

The subtraction method for removing the power-line (PL) interference from ECG signals [1, 8] shows high efficiency when the sampling sate (SR) Φ and the PL frequency F are synchronized, i.e. when $\Phi / F = n$ is integer. The method includes three main phases (the following interpretation is for odd value of n):

1. Each current sample is checked whether it belongs to a linear segment of the ECG signal, normally contaminated by PL interference. The introduced criterion of linearity is $|D_i| \le M$, where

$$D_i = \left(X_{i-n} - X_i\right) - \left(X_i - X_{i+n}\right), \quad (1)$$

which is called further D-filter, represents a second difference between signal samples, located each other at distance of one or more periods of the PL frequency, thus eliminating the interference influence. M is a practically chosen value of constant or variable threshold. Theoretically, the acceleration of the linear part of the signal should be zero, but ideal straight lines can not be found in a physiological signal.

2. The PL interference is removed by a low-pass filter type 'moving average', which is called K-filter. The free of interference sample Y_i is obtained by

$$Y_i = \frac{1}{n} \sum_{j=-(n-1)/2}^{(n-1)/2} X_{i+j} = \frac{1}{n} \left(\sum_{j=-(n-1)/2}^{-1} X_{i+j} + X_i + \sum_{j=1}^{(n-1)/2} X_{i+j} \right). \quad (2)$$

Simultaneously, a simple subtraction of filtered from non-filtered samples gives the value of the current interference B_i, which is stored in a temporary buffer.

$$B_i = X_i - Y_i = \frac{1}{n}\left(- \sum_{j=-(n-1)/2}^{-1} X_{i+j} + (n-1) X_i - \frac{1}{n} \sum_{j=1}^{(n-1)/2} X_{i+j} \right) \quad (3)$$

3. If the current sample does not belong to linear segment, a preceding value that corresponds to the same interference phase is taken from the buffer $B_{i-1}, B_{i-2}, \ldots, B_{i-n}$

$$B_i = B_{i-n}\,. \quad (4)$$

It is subtracted from the contaminated sample

$$Y_i = X_i - B_i \quad (5)$$

and is stored back at the current position in the buffer.

A more sophisticated approach to the subtraction method [4] includes the case of non-multiplicity between Φ and *F*, i.e. when Φ/*F* is a real number. The sample to be subtracted is determined by a larger number of additionally filtered values of the temporary buffer. If moving average is applied in case of multiplicity on *n* consecutive samples and the middle value is calculated by $B_{i-(n-1)/2} = -\sum_{j=-n+1}^{-(n-1)/2-1} B_{i+j} - \sum_{j=-(n-1)/2+1}^{0} B_{i+j}$, the obtained filter, called B-filter, allows the computation (extrapolation) of B_i by

$$B_i = -\sum_{j=-n+1}^{-(n-1)/2-1} B_{i+j} - B_{i-(n-1)/2} - \sum_{j=-(n-1)/2+1}^{-1} B_{i+j} = -\sum_{j=-n+1}^{-1} B_{i+j}\,. \quad (6)$$

Equation (6) is proposed to be used instead of equation (4).

In a first version of the subtraction method, the PL frequency variations are compensated by hardware synchronization of the analog-to-digital converter (ADC) to a given interference phase. As the PL frequency is *n* times lower than the SR, synchronization is applied at every *n*-th sample only, while the time-intervals between the other samples are generated by an internal timer.

The errors of such type of synchronization are analyzed in [6]. The authors report for error elimination by continuous measurement of the period of the PL frequency, which is divided by *n* to determine inter-sample intervals corresponding to the PL variations.

Hardware measurement of the PL period is practically impossible in battery powered devices. A software solution for measuring the interference frequency throughout the ECG signal subjected to band-pass filtering from 49 to 51 Hz is proposed in [9]. The PL period is calculated using amplitudes of two consecutive samples located bellow and above the zero line.

Both hardware and software determination of the PL period require an embarrassing to some extent hardware ADC synchronization. Besides, the time-depended parameters have to be recalculated to meet the standards for accuracy [2] especially in large deviation of PL frequency [3]. Another disadvantage of the hardware synchronization is that it cannot be applied on already recorded signals.

The theoretic base of applying the subtraction method to the general case of no multiplicity between SR and Φ is consecutively developed in [4, 5 and 10]. The expressions of the K- and B-filters are modified using auxiliary high-pass filters. The K-filter presented by equation (2) is transformed in a high-pass filter, called (1-K)-filter.

2. Theoretic study of the PL frequency deviations around its rated value

The variation of the PL frequency represents a specific case of non-multiplicity. Therefore, we applied modified filters obtained by using the low-pass K-filter from equation (2) as an auxiliary filter. The new filters are given by:

$$D_i^* = D_i - \left(X_i - Y_i\right)\frac{D_F}{1-K_F}; \tag{7}$$

$$Y_i^* = Y_i - \left(X_i - Y_i\right)\frac{K_F}{1-K_F} = X_i - \frac{X_i - Y_i}{1-K_F}; \tag{8}$$

Where D_F and K_F are transfer coefficients of the corresponding filters for the PL frequency. The adapted B-filter becomes $B^*_{i-(n-1)/2} = B_{i-(n-1)/2}/(1-K_F)$.

The modified algorithm of the subtraction method uses D_i and Y_i, which are first computed from (1) and (2) and then recomputed by means of (7) and (8). Taking in consideration the adapted B-filter, the extrapolation equation (6) for nonlinear sections is transformed in:

$$B_i^* = -\sum_{j=-n+1}^{-(n+1)/2} B^*_{i+j} - \left[1-nK_F\right]B^*_{i-(n-1)/2} - \sum_{j=-(n-3)/2}^{-1} B^*_{i+j}\,. \tag{9}$$

The coefficients D_F and K_F are dynamically recalculated to compensate the PL frequency variations during the interference removal procedure. Let the power-line frequency changes with *dF*. Then, according to equation (9), the coefficient K_{F+dF} has to be computed by

$$K_{F+dF} = \left(1 + \frac{\sum\limits_{j=-n+1}^{-(n+1)/2} B^*_{i+j} + \sum\limits_{j=-(n-3)/2}^{-1} B^*_{i+j} + B^*_i}{B^*_{i-(n-1)/2}}\right)\frac{1}{n}\,. \tag{10}$$

The coefficient D_{F+dF} is calculated using established relation for $D_{F+dF} = f(K_{F+dF})$. The new values of D_{F+dF} and K_{F+dF} are substituted for D_F and K_F in equations (7), (8) and (9) for processing the next sample.

Further, K_{F+dF} is used for calculating the diversion of the PL frequency. Assuming some transformations over equation $K_{F+dF} = \frac{1}{n}\cdot\frac{\sin\frac{n\pi(F+dF)}{\Phi}}{\sin\frac{\pi(F+dF)}{\Phi}}$, we may derive

$$dF \approx -K_{F+dF} \frac{\Phi}{\pi} \sin \frac{\pi F}{\Phi}. \qquad (11)$$

The relation $D_{F+dF} = f(K_{F+dF})$ becomes

$$D_{F+dF} \approx -4n^2 K_{F+dF}^2 \sin^2 \frac{\pi F}{\Phi}. \qquad (12)$$

The flow chart of the procedure for compensating the PL frequency variations is shown in fig.1. Some restrictions are introduced when the program has been written:

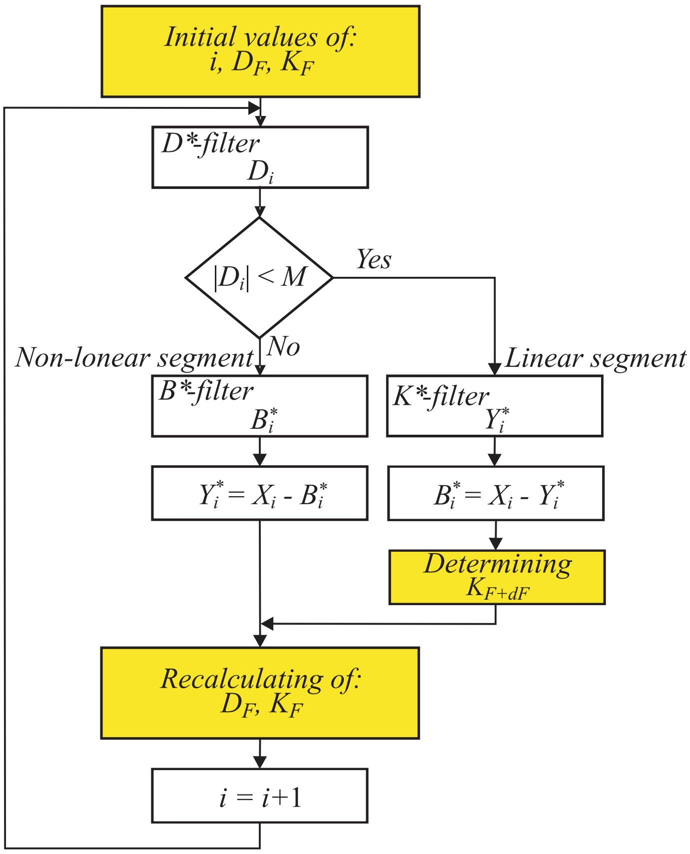

Fig.1. Flow chart of the subtraction method with compensation of the PL frequency deviation.

1. The K_{F+dF} calculation is protected against division by zero. The division is not performed if $B^*_{i-(n-1)/2}$ is less than one bit;

2. The coefficient K_{F+dF} is restricted within the allowed range of PL frequency variation;

3. The maximum speed of K_{F+dF} alteration is limited;

4. Since the PL frequency compensation is organized as a typical feedback control, the instability is avoided by a proportionally integral rule of adjustment.

$$K_F = \frac{2n-1}{2n} K_F + \frac{1}{2n} K_{F+dF}.$$

3. Experimental Results

The results of testing a real ECG signal with sampling rate Φ = 250 *Hz* and emulated PL frequency F = 50 *Hz* (n=5) is shown in figure 3. The initial values of the coefficients are $D_F = 0$ and $K_F = 0$.

An abrupt change in PL frequency from 52 to 48 *Hz* is simulated in the middle of the epoch. The last trace in fig. 2a shows the PL frequency (curve *a*), the estimated according to equation (11) diversion of the PL frequency (curve *b*) and the criterion for linearity (curve *c*). Correction in the D-filter is not introduced, because of the small influence of the PL frequency deviation. The time necessary to be reached the stationary value of the changed PL frequency (fig. 2*a*) is about 0.5 s. The error committed is less than 20 μV.

The result without PL frequency compensation is shown for comparison in fig. 2b. The error is more than ten times higher, especially within the QRS complexes.

Fig. 2c presents the case of a gradually frequency change within 1 s. It can be seen that the compensation process falls behind, because of erroneously detected non-linear segments. Obviously, a dynamic criterion for linearity [7] is suitable to be used in such case. The time for overtaking the new stationary frequency remains the same, but the transient between the two stationary frequencies is accompanying by a lot of non-linear segments detected. This is due to the specifics of the algorithm, which compensates frequency changes during linear segments only. It is possible to speed up the algorithm, if a compensation step can be extrapolated in non-linear segments.

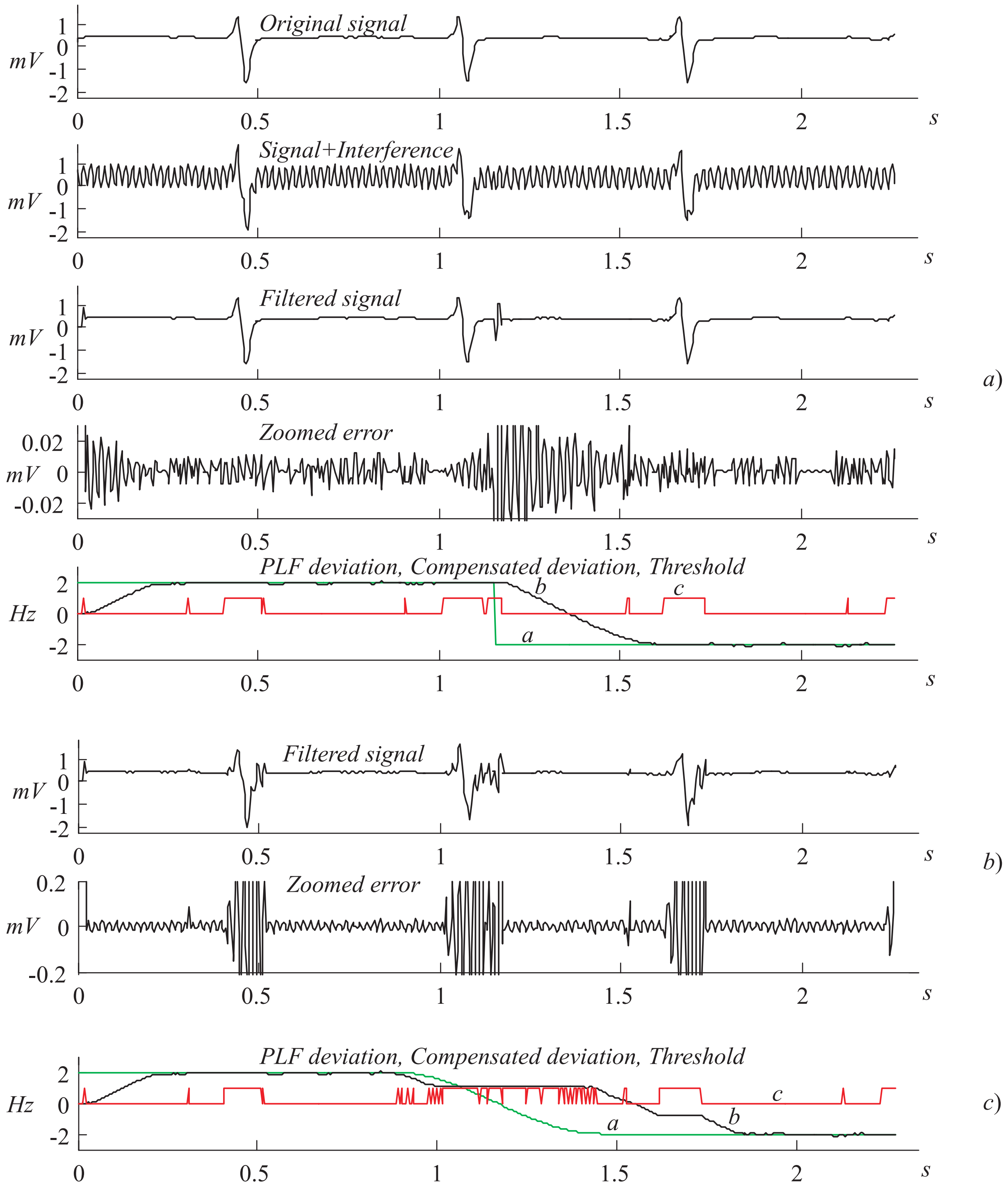

Fig.2. Experiment with automatic compensation of the PL frequency variation in real ECG signal.

In case of even multiplicity ($n=2m$), equation (2) should be changed to

$Y_i = \frac{1}{n}\left[\sum_{j=-n/2+1}^{n/2-1} X_{i+j} + \frac{X_{i-n/2}+X_{i+n/2}}{2}\right]$ that will introduce modification of equations (9), (10), (11) and (12) as $B_i^* = -B_{i-n}^* - 2\sum_{j=-n+1}^{-n/2-1} B_{i+j}^* - 2\left(1-nK_F\right)B_{i\text{-}n/2}^* - 2\sum_{j=-n/2+1}^{-1} B_{i+j}^*$ for (9),

$$K_{F+dF} = \left(1 + \frac{\frac{B_{i-n}^*}{2} + \sum_{j=-n+1}^{-n/2-1} B_{i+j}^* + \sum_{j=-n/2+1}^{-1} B_{i+j}^* + \frac{B_i^*}{2}}{B_{i\text{-}n/2}^*}\right)\frac{1}{n}$$ for (10), $dF \approx -K_{F+dF}\frac{\Phi}{\pi}\mathrm{tg}\frac{\pi F}{\Phi}$

for (11) and $D_{F+dF} \approx -4n^2 B_{F+dF}^2 \,\mathrm{tg}^2 \frac{\pi F}{\Phi}$ for (12). The other equations as well as the algorithm remain the same.

4. CONCLUSIONS

A modification of the subtraction method for interference removing from ECG signals is elaborated to compensate the PL frequency variations, which are considered as dynamically appearance of non-multiplicity between sampling rate and PL frequency. Appropriate formulae for correction of basic filter coefficients are derived. Experiments with odd multiplicity (n=2m+1) show that the method successfully compensates the PL frequency variations, except for the transients of gradually changes. The formulae for even multiplicity (n=2m) are derived too.

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