Biomedical Signal Processing and Control 3 (2008) 334–340

Contents lists available at ScienceDirect

Biomedical Signal Processing and Control

journal homepage: www.elsevier.com/locate/bspc

Short communication

Power-line interference elimination from ECG in case of non-multiplicity between the sampling rate and the power-line frequency

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ARTICLE INFO

Article history:
Received 17 August 2007
Received in revised form 4 March 2008
Accepted 21 April 2008
Available online 20 June 2008

Keywords:
ECG
Power-line interference
Interference elimination
Subtraction procedure

ABSTRACT

The paper deals with some aspects of the subtraction procedure, which removes the power-line interference (PLI) without affecting the components intrinsic to ECG. This procedure is based on the following principles: the interference is cancelled in linearly going ECG segments that have near to zero frequency content using moving averaging; the extracted samples are saved in a buffer and are then subtracted from the remaining parts of the signals. The accuracy of the subtraction procedure is analysed and improved in the cases of non-multiplicity between the sampling rate and the rated interference frequency. Extrapolation filters are applied over the buffer samples. Experiments with synthesised and real signals are carried out to assess the filter's stability. The results obtained show that the improved subtraction procedure removes the PL interference from ECG signals regardless of the type of multiplicity, odd or even, between the sampling rate and the power-line frequency.

1. Introduction

ECG recordings are often contaminated by residual power-line (PL) interference despite the very high common mode rejection ratio of the biomedical amplifiers. This is due to differences in the electrode impedances and to stray currents through the patient and the cables, resulting in transformation of the common mode voltage into a false differential signal [1,2]. The residual PL interference (PLI) may corrupt the proper function of automatic ECG analysis, which presumes correct delineation of ECG wave boundaries [3].

Sophisticated but conceptually traditional digital filters are known to suppress to different extent the components intrinsic to ECG around the PL frequency [4–6] and/or to introduce significant distortion in the QRS and ST-segment portions because of filter ringing and extended conversion time [7–11].

These drawbacks have been overcome some decades ago by the so called subtraction procedure [12]. Its principle consists of: (i) applying linear phase digital filter (moving averaging) on linearly going ECG segments that have near to zero frequency content (usually physiological baseline, low amplitude P-waves and some small parts of T-waves); (ii) continuous updating and memorizing the removed phase locked PLI components B_i; (iii) subsequent subtracting the corresponding component from the contaminated input ECG sample X_i wherever non-linear segments are encountered, thus obtaining a free of interference output ECG sample $Y_i = X_i - B_i$. Later on, many improvements of the procedure have been introduced to cope with PL amplitude and frequency variations, tremor contamination as well as the non-multiple sampling, which lead to a real (non-integer) number n of samples within one rated PL period [13–17]. This is the case, for example, with some AHA database recordings, which are digitised with sampling rate $\Phi = 250$ Hz but contain interference with PL frequency $F = 60$ Hz.

The problem was overcome by the generalised structure of the subtraction procedure [15,16] shown in Fig. 1. Three main modules are present: *linear segment detection*, *interference extraction* and *interference temporal buffer*. Basic manipulations are formulated as filters. *Linear segment detection* is checked by the so called *D-filter* with zeros in $f = 0$ and $f = F$. The most used version of *D-filter* in the case of multiplicity is given by $D_i = X_{i-n} - 2X_i + X_{i+n}$, which corresponds to the signal acceleration. The *interference extraction* is done by subtracting the result of a *K-filter* from the contaminated current sample X_i. This filter cancels the interference by zero in $f = F$ and unity gain in $f = 0$ in its frequency response thus implementing the moving averaging within window equal to the interference period.

When Φ and F are multiple, the phase locked interference B_i has practically the B_{i-n} value. Otherwise, a phase difference appears

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1746-8094/$ – see front matter
doi:10.1016/j.bspc.2008.04.006

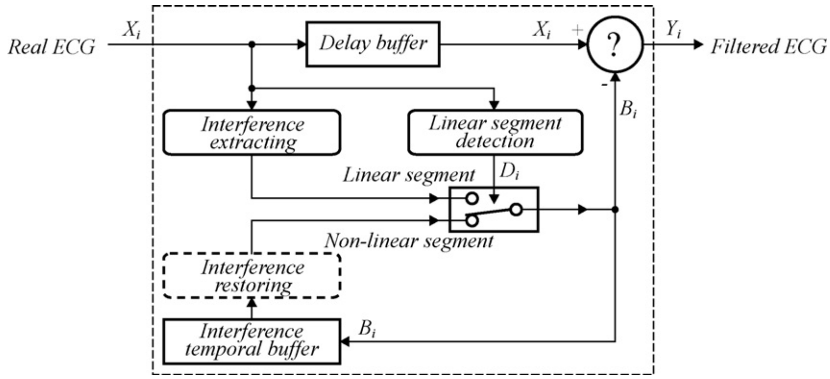

Fig. 1. Generalised structure of the subtraction procedure.

between them introducing amplitude error, as illustrated in Fig. 4. Therefore, B_i is estimated by means of the temporal buffer containing the filtered samples $B_{i-1}, B_{i-2}, \ldots, B_{i-n^*}$. Here n^* stands for the truncated real number n. To compensate for the error, another module *interference restoring* is included in the structure. The above-mentioned filters are modified [15]. A third extrapolation *B-filter* with linear phase response and unity gain for $f = F$ is introduced specifically for the non-multiplicity. It is usually synthesised from the *K-filter*, and is used for modifying the sample B_i before its subtraction from the signal during non-linear segments.

2. Aim of the study

This study was challenged by colleagues from the Cairo University [18], who experimented, according to the equations given in Ref. [16], PLI removal in the case of even non-multiplicity with high truncated real number n^* obtained with Φ = 500 Hz and F = 41 Hz (see *experiment* 5, Fig. 13, below).

It was found that in such circumstances the subtraction procedure is prone to auto-excitation because the used *B-filter* reproduces the structure of the even *K-filter* $Y_i = (1/n^*)\sum_{j=-n^*/2+1}^{n^*/2-1} X_{i+j} + (X_{i-n^*} + X_{i+n^*})/2$, which includes an additional n^*+1-th sample to the even number of n^*. (This sample is necessary for measuring the amplitude difference between two samples with the same interference phase [15,16] used further for correcting and shifting the averaged virtual middle sample into one of the surrounding real sample positions before calculating the corresponding phase locked interference.)

In this way, the vector coefficient of the sample that has to be compensated by the *B-filter* becomes half of the other buffer coefficients. Therefore, auto-excitation may occur in the case of high even n^* if the buffer samples contain residual low frequency components and/or the non-linear segment is long enough to accumulate computing error.

The aim of this study is to improve the accuracy of the subtraction procedure for particular cases of high truncated even number n^* by a modified evaluation of the buffer content. The *B-filter* is synthesised using reorganised *K-filter* (marked as K_B-*filter*), which includes more precisely calculated virtual middle buffer sample.

3. Theory, method and algorithm

This approach uses common K_B-*filter* for odd and even n^* samples with rectangle impulse response to estimate B_i by the buffer content. The filter has transfer coefficient K_{BF} for $f = F$. The corresponding equations are:

$$\frac{1}{n^*}\sum_{j=-n^*+1}^{0} B_{i+j} = B_{\text{mid}} K_{\text{BF}}; \tag{1}$$

$$K_{\text{BF}} = \frac{1}{n^*}\cdot\frac{\sin(n^*\pi F/\Phi)}{\sin(\pi F/\Phi)}; \tag{2}$$

$$B_i = B_{\text{mid}} n^* K_{\text{BF}} - \sum_{j=-n^*+1}^{-1} B_{i+j}. \tag{3}$$

Here B_{mid} is the middle term of the filter area $B_i, B_{i-1}, \ldots, B_{i-n^*+1}$. The K_B-*filter* $\mathbf{K}_B \equiv (1/n^*)\sum_{j=-n^*+1}^{0} B_{i+j}$ is low pass type with coefficient vector $\vec{\mathbf{K}}_B$ consisting, in this case, of either odd or even n^* terms with equal weight coefficient $1/n^*$. The filter is further transformed in high pass type *B-filter* [15,16] by subtracting $\mathbf{K}_B$ from B_{mid}, e.g. $\mathbf{B} \equiv B_{\text{mid}} - (1/n^*)\sum_{j=-n^*+1}^{0} B_{i+j}$. It has $1 - K_{\text{BF}}$ transfer coefficient at $f = F$. The next modification, the B^*-*filter*, is expressed by $\mathbf{B}^* \equiv \mathbf{B}/(1 - K_{\text{BF}})$ or $\mathbf{B}^* \equiv -[(1/n^*)\sum_{j=-n^*+1}^{0} B_{i+j} + B_{\text{mid}}]/(1 - K_{\text{BF}})$ with transfer coefficient equal to 1 at $f = F$, which leads implicitly back to the Eq. (1).

If the K_B-*filter* is applied once more on the shifted left by one content of the filter area $B_{i-1}, B_{i-2}, \ldots, B_{i-n^*}$, now with middle sample $B_{\text{mid}-1}$ (see Fig. 4), one may write $B_{i-n^*} = B_{\text{mid}-1} n^* K_{\text{BF}} - \sum_{j=-n^*+1}^{-1} B_{i+j}$. By substituting the sum $\sum_{j=-n^*+1}^{-1} B_{i+j} = B_{\text{mid}-1} n^* K_{\text{BF}} - B_{i-n^*}$ in Eq. (3), a simplified formula for calculating B_i is obtained

$$B_i = B_{i-n^*} + n^* K_{\text{BF}}(B_{\text{mid}} - B_{\text{mid}-1}). \tag{3a}$$

When n^* is odd, B_{mid} and $B_{\text{mid}-1}$ coincide in time with the real samples $B_{i-(n-1)/2}$ and $B_{i-(n+1)/2}$ and Eq. (3a) becomes

$$B_i = B_{i-n^*} + n^* K_{\text{BF}}(B_{i-(n-1)/2} - B_{i-(n+1)/2}). \tag{4}$$

Fig. 2 represents the B^*-*filter* synthesis for PLI evaluation in the case of Φ = 250 Hz, and F equal to 48 and 52 Hz, both of them with odd number n^* = 5. The traces are obtained in MATLAB environment with the filter vector coefficients: $\vec{\mathbf{K}}_B = [1 \quad 1 \quad 1 \quad 1 \quad 1]/5$; $\vec{\mathbf{B}} = [-1 \quad -1 \quad 4 \quad -1 \quad -1]/5$; $\vec{\mathbf{B}}^* = [-1/5 \quad -1/5 \quad 4/5 \quad -1/5 \quad -1/5]/(1 - K_{\text{BF}})$. K_{BF} is equal to -0.0412 for F = 48 Hz and to 0.0442 for F = 52 Hz.

The removal of PLI with F = 52 Hz when the sample B_i is calculated according to Eq. (4) is shown in Fig. 3. The first trace is original conditionally clean ECG signal. It is mixed by synthesised interference (second trace). The third diagram contains the clean signal together with the linearity criterion (0 in linear segments, 1 otherwise). The error (the difference between the conditionally clean signal and the processed signal) shown in the upper trace of