Adaptive Filtering of ECG Baseline Wander

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Abstract - **Baseline wander removal is a classical problem in ECG signal processing. We present a cascade adaptive filter to remove the baseline wander in the ECG preserving the overlapped deterministic low frequency components of the ECG, such as ST segment components. This cascade adaptive filter works in two stages. The first stage is an adaptive notch filter at zero frequency. The second stage is an adaptive impulse correlated filter that estimates the ECG signal correlated with the QRS occurrence. In both stages the LMS algorithm is used with different gain constants μ_1 and μ_2. We analyse the frequency response of the filter as a function of the μ_1 and μ_2 parameters, selecting those more appropriated for baseline removal. Finally, the performance of the filter is studied on an actual ECG affected by baseline drift.**

I. INTRODUCTION

Baseline wander in the ECG records produces artifactual data when measuring ECG parameters. These baseline interferences can be induced by electrode changes due to perspiration, movement and respiration, and are specially relevant in ECG records obtained during exercice testing. The frequency components of the baseline wander are usually below 0.5 Hz but, in case of stress test, this limit can be higher. Thus, these components can be in the same range than the low frequency ECG components like those of ST segment. Then, removal of the baseline can adversely alter the ECG clinical information.

Several methods have been used in the literature to eliminate the baseline wander. The more classical is ensemble averaging. This approach is adequated when the ECG signal remains constant in each beat, but this is not the situation in many actual ECG records. Other method is polynomial interpolation. Linear interpolation introduces significant distortions. A third order approximation called cubic spline [1] achieves better results. Interpolation makes use of a previous knowledge of the ECG isoelectric levels estimated from the PR intervals, also called knots. This is a nonlinear approach which performance depends on the knots determination accuracy, and it is degraded as the knots become more separated (low heart rate). Other method that overcomes this problem is digital narrow-band linear-phase filtering [2]. This method can be implemented in real time, but has two major problems: first, the filter requires to be a FIR filter with a long impulse response and a large number of coefficients; second, given that ECG and baseline wander spectra usually overlap, it is not possible to remove the baseline wander with a linear filter without distorting the ECG components. Sörnmo [3] has proposed a time-varying filtering technique that selects different cut-off frequencies of the linear filter as a function of the heart rate or the baseline level. This filter improves the time invariant FIR filter performance, but can yet distort the ST components and has high computational requirements. Adaptive filtering has been recently proposed [4] to cancel the baseline drift. This filter is an adaptive transversal filter with one weight, where the reference input is a constant with a value of 1 and the primary input is the ECG signal [5]. This filter, using the LMS algorithm in the adaptation process, is equivalent to a linear notch filter, that takes the advantage of the adaptive implementation but, still modifies the ST components [4].

The AHA recommends that the cut-off frequency of a high-pass filter that removes the dc component of the ECG should not exceed 0.05 Hz [6] and in [2] is stated that, if linear phase is preserved, the cut-off frequecy can be chosen as the fundamental frecuency of the heart rate or lower (< 0.8 Hz). Thus, if these conditions are satisfied baseline wander of higher frequency will remain at the filtered ECG. To solve this problem we propose a cascade adaptive filter that first filter with the high-pass notch filter [4] at a cut-off frequency lower than the 0.8 Hz, removing dc component and frequencies lower than those where ST components appears. After that, taking advantage of the recurrent behaviour of the ECG signal, we apply an adaptive impulse correlated filter (AICF) [7] to remove remaining baseline wander preserving the ST components.

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II. THE CASCADE ADAPTIVE FILTER

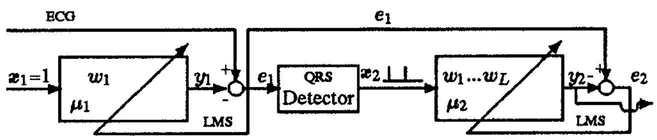

Figure 1: The cascade adaptive filter

The cascade adaptive filter that we propose is plotted in Fig. 1. The first stage is an adaptive transversal high-pass filter [5] with one weight that uses the LMS algorithm ($w_{1,k+1} = w_{1,k} + 2\mu_1 e_1 x_1$). The primary input is the ECG signal contaminated by baseline wander. The reference input (x_1) is a constant of value 1 and the output signal is the error signal e_1. The cut-off frequency at -3 dB is $f_c = \frac{\mu_1}{\pi} f_s$, where f_s is the sampling rate. The convergence time is $1/(4\mu_1)$ samples. Selecting $\mu_1 = 0.001$ and a $f_s = 1000$ Hz, we get a convergence time of 0.25 seconds and a cut-off frequency of 0.3 Hz. This frequency satisfies the limit stated in [2] (0.8 Hz) but in the output signal e_1 still remain baseline contaminations of frequencies higher than 0.3 Hz.

To remove this remaining contamination we filter again the signal e_1 with an AICF [7]. This second stage adaptive filter uses as primary input the signal we want to filter e_1, and as reference input a unit impulse sequence (x_2) correlated with each QRS complex. This needs a QRS detector to generate the impulse sequence (Fig. 1). The filter requires a number L of weights as large as the number of samples that the P-QRS-T complex spans. The output signal is taken at y_2. The gain constant of the LMS algorithm in this second stage filter

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is μ_2 that implies a convergence time of $L/(4\mu_2)$ samples. To study the behaviour of this AICF we can suppose the primary input as the consecutive linking of the L sample recurrences of each P-QRS-T complex, and the reference input as a periodic unit impulse each L samples. In [8] we have shown that this filter, when using the LMS algorithm, is equivalent to a linear exponentially weighted averager which impulse response takes the form

$$h_2[n] = \sum_{m=1}^{N} 2\mu_2(1-2\mu_2)^{N-m}\delta[n-mL] \quad (1)$$

where N is the number of processed recurrences. Calculating the discrete Fourier transform $H_2(\Omega)$ of $h[n]$ when $N \to \infty$ we have, except by a time delay factor, that

$$H_2(\Omega) = \frac{2\mu_2}{L}\sum_{k=0}^{L-1}\frac{1}{1-e^{[1/(\tau L)\ -\ j(\Omega-2\pi k/L)]}} \quad (2)$$

where $e^{-1/\tau} = (1-2\mu_2)$ and Ω is the normalized frequency. This filter is a comb filter with lobes repeated at frequencies multiples of the fundamental frequency (f_0) of the P-QRS-T complex $f_0 = f_s/L$. The -3 dB cut-off of each lobe is $\frac{\mu_2}{\pi}f_0$ far from the central frequency of the lobe. Then this filter extracts the deterministic signal that repeats at each QRS recurrence including the dc component (k=0 in (2)). This dc component is not desirable since it includes baseline variations, but if μ_2 is selected to give $\frac{\mu_2}{\pi}f_0 < \frac{\mu_1}{\pi}f_s$ the first lobe of the comb filter will be attenuated by the high-pass filter of the first stage. This condition implies that $\mu_2 < \mu_1 L$. The resulting cascade filter removes the dc component and those components not correlated with the QRS complex (remaining baseline wander), preserving the QRS correlated components (ST segment). In Fig. 2 we plot the resulting transfer

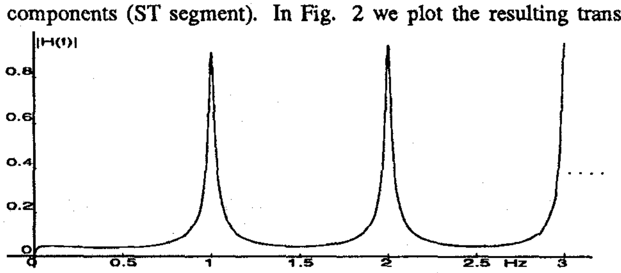

Figure 2: Transfer function of the cascade filter

function module $|H(f)|$ of the cascade filter for L=1000 samples, f_s = 1000 Hz (f_0 = 1 Hz), μ_1 = 0.001 and μ_2 = 0.05 ($\mu_2 < \mu_1 L$). This graphic must be interpreted carefully because the comb filter does not apply to the output of the high-pass filter e_1. The first stage high-pass filter applies to the ECG signal and the AICF applies to the output of the previous stage (e_1) but the performance study has been done assuming periodic inputs (removing the inter P-QRS-T complex interval). Thus, remaining baseline variations at frequencies multiples of f_0 in the e_1 signals, that in accordance with Fig. 2 would remain at the output y_2, really would be removed because they are not correlated with the QRS occurrence. The ST components at frequencies different of f_0 multiples, in the original ECG and in the e_1 signal, that in accordance with Fig. 2 would be removed, are preserved because are correlated with the QRS complex and then translated to multiples of f_0 when considering the comb filter.

III. APPLICATION TO ACTUAL ECG RECORDS

In Fig. 3 we present an actual ECG affected by baseline wander and sampled at 1000 Hz, the output of the high-pass filter, e_1, using μ_1=0.001 and the cascade filter output y_2 using μ_2=0.05. We can corroborate how the high-pass filter, with a cut-off frequency of 0.3 Hz, does not remove the whole baseline wander in e_1 and then the AICF, at the filter output y_2, removes the remaining variations together with all signal interferences not correlated with the QRS complex. This filter, in addition to removing baseline wander, attenuates other interferences present in the ECG as was already noted in [8].

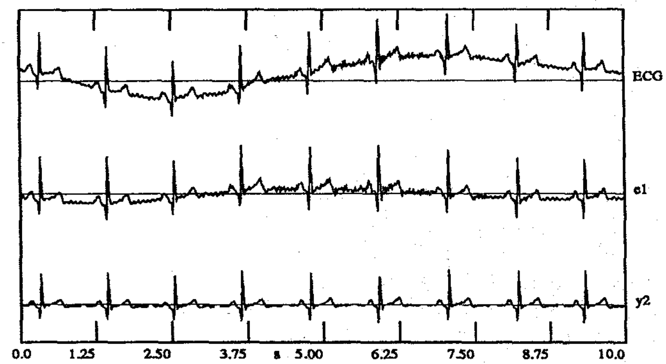

Figure 3: ECG, high-pass filtered and cascade filtered signal

IV. CONCLUSIONS

The proposed cascade adaptive filter allows to remove the baseline wander present in the ECG preserving the components of clinical information. It removes frequencies below 0.3 Hz and those higher than 0.3 Hz not correlated with the QRS occurrence. Then, as the ST segment repeats with each QRS it has frequency components that fall into the fundamental lobe and its armonics of Fig. 2, and then are preserved at the output filter. This filter can efficiently be implemented in the adaptive form with the only requirement of a real-time QRS detector and does not need to compute the knots as in the cubic splines technique. In addition [8] the filter removes all interferences not correlated with the QRS as 50/60 Hz, motion artifact and EMG.

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