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Use of power-line interference for adaptive motion artifact removal in biopotential measurements

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Abstract. Motion artifacts (MA) have long been a problem in biopotential measurements. Adaptive filtering is widely used for optimal noise removal in many biomedical applications. However, the existing adaptive filtering methods involve the use of additional sensors, limiting the applicability of adaptive filtering for MA reduction. In the present study, a novel adaptive filtering method without need for additional sensors is proposed. In biopotential measurements, movement of the electrodes and their leads may cause variations not only in the skin and half-cell potential (motion artifacts), but also in the electrode-skin impedance. Such impedance variations may also cause power-line interference modulation (PLIM), resulting in additional spectral components around the power-line interference (PLI) in the frequency domain. Demodulation of the PLI may reflect the movement-induced electrode-skin impedance variation, and can therefore represent a reference signal for the adaptive filter. Preliminary validation on ECG measurements with seven volunteers showed a high correlation coefficient ($R = 0.97$) between MA and PLIM, and excellent MA removal by the proposed adaptive filter, possibly leading to improved analysis of biopotential signals.

Keywords: Adaptive filtering, Motion artifacts, Biopotential measurements, Power-line interference, Demodulation.
1. Introduction

Surface electrodes, e.g., Ag/AgCl electrodes, are widely used for biopotential measurements, such as electrocardiography (ECG), electroencephalography (EEG), electromyography (EMG), and electrohysterography (EHG), in both clinical and research settings (Huigen et al. 2002, Niedermeyer & da Silva 2005, Xu et al. 2015, Rabotti & Mischi 2015). To sense any biopotential, a current flow from the body to the electronic measurement circuitry is necessary. Current is carried by ions in the body while it is carried by electrons in the electrode and its lead. An electrolytic gel, containing cations of the electrode metal and anions, is therefore added to the interface between the skin and the metallic electrode to allow for the electro-ionic conversion by a redox reaction, creating a net current across the interface (Webster 2009).

When a metal electrode comes into contact with a gel (electrolyte), a double layer of charge, known as half-cell potential, forms at the metal-electrolyte interface (Geddes 1972). Gel movement relative to the electrode can disturb the ionic distribution in this interface producing an interference in the half-cell potential (Kahn 1965). This interference is generally referred to as motion artifacts (MA). MA have long been a problem in biological signal measurements using contact electrodes and may, in fact, mask the desired signal or cause an abrupt shift in the baseline, resulting in misdiagnosis and, therefore, in delayed or inappropriate treatment decisions in clinical applications (Srikureja et al. 2000, Knight et al. 2001).

Variation of the half-cell potential, however, is not the only source contributing to the generation of MA. In addition to the metal-electrolyte interface, the electrolyte-skin interface can be another source of MA. As compared to the half-cell potential variation, higher MA may be caused by gel movement relative to the skin, and the highest artifacts may be skin potential variation due to the deformation of the skin beneath the electrode (Ödman 1981, Ödman & Öberg 1982, de Talhouet & Webster 1996, Tam & Webster 1977, Burbank & Webster 1978). As demonstrated by our previous study, MA induced by half-cell potential and skin potential variation can be considered as additive noise with respect to the desired signal (Xu et al. 2013).

A time varying electric field is usually present in the surrounding environment during biopotential measurements. This would cause no problem in measurements with perfect grounding and active shielding. However, active shielding is achieved by feeding the electrode signal at the inner wire back to the shield, which needs to be driven by a low impedance source. Therefore, additional hardware, e.g. a voltage follower, is required between the inner electrode signal and the shield, resulting in increased hardware complexity and, therefore, increased power consumption (Van Rijn et al. 1990). Optimal use of active shielding is therefore unsuitable for applications where low power consumption is of key priority, e.g., continuous ambulatory monitoring. Furthermore, perfect grounding is also not available in many ambulatory settings. In those cases, a power-line interference (PLI), at a frequency \( f_p \) equal to 50 Hz in Europe and 60 Hz in the United States, can be observed in the obtained recordings.
The PLI source appears at the input terminal of the amplifier (Huhta & Webster 1973, Simakov & Webster 2010). The amplified voltage is the product of the PLI source and the impedance difference between the two recording electrodes (Huhta & Webster 1973, Simakov & Webster 2010). If there are no movements in the electrodes and their leads, the difference in electrode impedance is constant and the PLI should appear at $f_p$ only. However, movement of the electrodes or the leads may cause variation in the electrode-skin impedance (Ödman 1981, de Talhouet & Webster 1996). Such impedance variation modulates the PLI, resulting in a shift of the PLI in the frequency domain, which may be referred to as PLI modulation (PLIM).

Several methods have been proposed in the literature to remove MA from biopotential recordings (Mischi & Cardinale 2009, Mischi et al. 2010, Izzetoglu et al. 2005, Goldstein et al. 1998, Chen et al. 2006). A band-stop filter was suggested to suppress MA from EMG data recorded during vibration exercise (Mischi & Cardinale 2009, Mischi et al. 2010). However, by this approach, some useful signals located within the stop bands may be removed. Wiener filtering has also been employed for MA reduction in some applications (Izzetoglu et al. 2005). However, the optimal use of a Wiener filter requires the statistics of the entire data (Chen et al. 2006). Moreover, a Wiener filter is not suitable for non-stationary signals (Goldstein et al. 1998).

Adaptive filtering has been widely employed for noise removal in many biomedical application areas (Widrow et al. 1975, Tong et al. 2002, Thakor & Zhu 1991, Xu et al. 2013, Thakor 1987, Luo & Tompkins 1995, Degen & Jackel 2008, Serteyn et al. 2015). An adaptive filter performs a decorrelation between the noise-contaminated signal and a clean reference signal; any noise correlated to the reference signal is removed from the original signal. The reference signal of an adaptive filter is usually measured by an additional sensor. An accelerometer is most commonly used to measure the movement of an electrode (Tong et al. 2002, Xu et al. 2013), while electrode-skin impedance is also monitored by injecting an external current, providing a reference for noise cancelation (Degen & Jackel 2008, Serteyn et al. 2015). However, the use of an accelerometer or the injection of an external current may increase the complexity of the recording system, limiting the application of adaptive filtering for MA reduction in several ambulatory conditions.

In this study, a novel adaptive filtering method is proposed for MA removal in biopotential measurements which does not require an additional sensor or any signal injection. As mentioned before, in biopotential measurements without perfect grounding and shielding, movement of the electrodes and their leads may cause not only MA, but also electrode-skin impedance variation, resulting in PLIM, observed as a shift of the PLI in the frequency domain. It is therefore reasonable to hypothesize that the PLIM and MA are highly correlated as they are generated by the same movement. Based on this hypothesis, a reference signal for adaptive MA removal can be extracted by demodulating the PLI. After testing the relationship between MA and PLIM, we validated the proposed adaptive filter method by ECG measurements performed on seven subjects. To simulate MA, a dedicated vibratory setup was employed to generate
sinusoidal vibrations at different intensities (Xu et al. 2013).

After introducing the relevant background on PLI and PLIM in Section 2.1, the proposed adaptive filter is provided in Section 2.2. The experimental measurement and evaluation of the proposed method is then described in Section 2.3 and 2.4, respectively. Our results are reported in section 3, followed by some discussion (Section 4) and our conclusion (Section 5).

2. Method

2.1. Background: motion artifacts and power-line interference

In general, biopotentials are usually recorded in the presence of movement, such as respiration and body movement. Respiration occurs mainly in the range below 1 Hz (Sörnmo & Laguna 2005), while body movement occurs in the range from 1 to 10 Hz (Tam & Webster 1977, Burbank & Webster 1978). Respiration and body movement can cause variations in the half-cell potential and skin potential, resulting in MA, which have the same frequency as the original movement (<10 Hz). Furthermore, movement of the electrodes and their leads can also cause variation in the electrode-skin impedance (Ödman 1981, de Talhouet & Webster 1996), resulting in PLIM. The mechanisms involved in the generation of PLIM are described hereafter.

2.1.1. Displacement currents into the leads

During biopotential measurements, changes in the electric field intensity in the surrounding will capacitively couple a displacement current \( i_w(t) \) into the leads of the recording system. Here, \( i_w(t) \) is given as

\[
i_w(t) = A_w \sin(2\pi f_p t + \varphi_w),
\]

where \( f_p \) is the power-line frequency, and \( A_w \) and \( \varphi_w \) are the amplitude and phase of \( i_w \), respectively. Fig. 1 shows a representative recording situation for ECG measurements, in which the electrode leads are assumed to be unshielded and run close to each other. The displacement currents into each lead should therefore be approximately equal (Huhta & Webster 1973). Due to the high input impedance of the amplifier, any current coupled into the electrode leads flows through the associated electrode-skin impedance, \( Z_1 \) or \( Z_2 \), and the ground impedance \( Z_G \). The differential input between A and B, \( v_d(t) \), will be amplified. By assuming the internal body resistance to be negligible, we have

\[
v_A(t) = i_w(t)Z_1 + 2 \cdot i_w(t)Z_G,
\]

\[
v_B(t) = i_w(t)Z_2 + 2 \cdot i_w(t)Z_G,
\]

\[
v_d(t) = (Z_1 - Z_2)i_w(t) = Z_d A_w \sin(2\pi f_p t + \varphi_w),
\]

with \( Z_d = Z_1 - Z_2 \). If \( Z_1 = Z_2 \), the amplified voltage will be zero. An impedance difference between the two recording electrodes is often present and may reach 20 k\( \Omega \).
Values measured for 3-m cables show that $A_w \approx 6 \text{nA}$ (Huhta & Webster 1973). Therefore,

$$v_d(t) = (20 \text{k}$\Omega$) \cdot (6 \text{nA}) \cdot \sin(2\pi f t + \varphi_w) = (120 \mu \text{V}) \cdot \sin(2\pi f_p t + \varphi_w),$$

which would be an substantial level of interference as compared to the amplitude of the acquired ECG signals, in particular, fetal ECG (Vullings et al. 2013).

Without electrode or lead movement, $Z_d$ is constant and the interference should only be present at $f_p$. However, movement of the electrodes or leads may occur in real measurements, resulting in variation in the electrode-skin impedance (Ödman 1981, de Talhouet & Webster 1996). Assuming a single, sinusoidal movement for the electrodes and leads, the movement-induced time-varying electrode impedance $Z'_1(t)$ and $Z'_2(t)$ are given as

$$Z'_1(t) = Z_1 + A_1 \sin(2\pi f_m t + \varphi_1),$$
$$Z'_2(t) = Z_2 + A_2 \sin(2\pi f_m t + \varphi_2),$$

where $f_m$ is the frequency of the movement, $Z_1$ and $Z_2$ the constant impedances, $A_1$ and $A_2$ the AC impedance amplitudes, and $\varphi_1$ and $\varphi_2$ the phase of $Z'_1(t)$ and $Z'_2(t)$, respectively. As a result, the impedance difference between the two recording electrodes is no longer constant. The time-varying impedance difference, $Z'_d(t)$, can be expressed as

$$Z'_d(t) = Z_d + [A_1 \sin(2\pi f_m t + \varphi_1) - A_2 \sin(2\pi f_m t + \varphi_2)]$$

$$= Z_d + A \cos(2\pi f_m t + \frac{\varphi_1 + \varphi_2}{2}),$$

where

$$A = \sqrt{A_1^2 + A_2^2 - 2A_1 A_2 \cos(\varphi_1 - \varphi_2)}.$$  

Replacing $Z_d$ in (4) with $Z'_d(t)$ yields

$$v_d(t) = [Z_d + A \cos(2\pi f_m t + \frac{\varphi_1 + \varphi_2}{2})] A_w \sin(2\pi f_p t + \varphi_w)$$

$$= Z_d A_w \sin(2\pi f_p t + \varphi_w) + \frac{AA_w}{2} \sin[2\pi (f_p + f_m) t + (\varphi_w + \frac{\varphi_1 + \varphi_2}{2})]$$

$$+ \frac{AA_w}{2} \sin[2\pi (f_p - f_m) t + (\varphi_w - \frac{\varphi_1 + \varphi_2}{2})].$$

Therefore, two spectral components located at $f_p - f_m$ and $f_p + f_m$ can be observed and may be referred to as PLIM. Figure 2 shows an example illustrating in the frequency domain the MA and the PLI together with the PLIM.

2.1.2. Displacement currents into the body

The surface of the human body is subject to capacitive coupling and a displacement current $i_b(t)$ can flow through the body to the ground. Here, $i_b(t)$ is given as

$$i_b(t) = A_b \sin(2\pi f_p t + \varphi_b),$$

where $f_p$ is the power-line frequency, and $A_b$ and $\varphi_b$ are the amplitude and phase of $i_b$, respectively. This current brings the body to a potential that is different from the ground.
potential and is referred to as common mode potential, \( v_{cm}(t) \). This is determined by the displacement current \( i_b(t) \) and the ground electrode impedance \( Z_G \), as shown in Fig. 3.

By neglecting the internal body resistance and the current returning to the ground through the input impedance of the amplifier \( Z_{in} \), \( v_{cm}(t) \) can be calculated as

\[
v_{cm}(t) = Z_G \cdot i_b(t) = Z_G \cdot A_b \sin(2\pi f_p t + \varphi_b). \tag{12}
\]

Substituting typical values (\( A_b = 0.2 \) µA, \( Z_G = 50 \) kΩ (Webster 2009)) yields an amplitude for \( v_{cm}(t) \) equal to 10 mV. In poor electrical environments where \( A_b > 1 \) µA, the amplitude of \( v_{cm}(t) \) can be larger than 50 mV. For an ideal amplifier, this would cause no problem due to its infinite common mode rejection ratio (CMRR). However, real amplifiers have a finite CMRR and input impedance \( Z_{in} \). Therefore, the common mode potential \( v_{cm}(t) \) appears at the differential input of the amplifier through the voltage dividers (see Fig. 3):

\[
v_A(t) = v_{cm}(t) \cdot \left( \frac{Z_{in}}{Z_{in} + Z_1} \right), \tag{13}
\]

\[
v_B(t) = v_{cm}(t) \cdot \left( \frac{Z_{in}}{Z_{in} + Z_2} \right), \tag{14}
\]

\[
v_A(t) - v_B(t) = v_{cm}(t) \cdot \left( \frac{Z_{in}}{Z_{in} + Z_1} - \frac{Z_{in}}{Z_{in} + Z_2} \right). \tag{15}
\]

Assuming \( Z_1 \) and \( Z_2 \) much smaller than \( Z_{in} \) and substituting \( v_{cm}(t) \) in (12),

\[
v_A(t) - v_B(t) \cong \frac{Z_2 - Z_1}{Z_{in}} \cdot v_{cm}(t) = \frac{Z_d}{Z_{in}} Z_G A_b \sin(2\pi f_p t + \varphi_b). \tag{16}
\]
Similar to Section 2.1.1, without electrode and lead movement, the difference between \( Z_1 \) and \( Z_2 \) is constant and the interference appears only at \( f_p \). In the presence of movement in the electrodes and leads, \( Z_d \) becomes time-varying, \( Z_d'(t) \), as expressed in (8), and PLIM can be observed around \( f_p \), as expressed in (10).

### 2.2. Adaptive filtering

An adaptive filter can remove any noise correlated to a given reference signal from a noise-contaminated signal. In this study, the PLIM are exploited to extract the reference signal for adaptively removing MA in ECG recordings. The scheme of the proposed adaptive filter is shown in Fig. 4, in which the ECG signal is assumed to be recorded without shielding or adequate grounding and, therefore, contains the desired clean ECG signal, the MA, and the PLI together with the PLIM.

![Figure 3](image_url) Displacement currents coupled into the surface of the body in an ECG recording system.

![Figure 4](image_url) Scheme of the proposed adaptive filter.

As shown in the lower branch in Fig. 4, in order to obtain the reference signal \( r[n] \), the recorded ECG signal is first band-pass filtered around \( f_p \) (\( f_p \pm 10 \text{ Hz} \)), since the movement-induced MA is mainly ranging below 10 Hz (Tam & Webster 1977, Burbank & Webster 1978). The band-pass filtered signal, \( b[n] \), is then demodulated to obtain a signal \( r[n] \) ranging from 0 to 10 Hz, reflecting the movement-induced variation in the electrode-skin impedance.

In the upper branch in Fig. 4, the recorded ECG signal is first low-pass filtered (<80 Hz) to remove high-frequency noise. For PLI removal, several methods are available (Martens et al. 2006). In the present study, a band-stop filter is employed.
The filtered signal, $x[n]$, contains both the desired clean ECG signal and MA, $m[n]$, as given by

$$x[n] = ECG + m[n].$$

(17)

The reference signal $r[n]$ is then filtered with an adaptive finite impulse response (FIR) filter $W[n]$ in order to obtain an estimate $\hat{m}[n]$ of MA. By subtracting $\hat{m}[n]$ from $x[n]$, the desired clean ECG signal $\hat{s}[n]$ can be obtained.

An adaptive normalized least mean square (NLMS) algorithm is employed to estimate and remove the movement-induced MA in $x[n]$ (Xu et al. 2013). Assuming the ECG signal and $m[n]$ to be uncorrelated, the adaptive filter is designed such that the squared error $|\hat{s}[n]|^2$ is minimized (Widrow et al. 1975). The NLMS algorithm is then given as

$$W_{NLMS,opt} = \arg \min_W |\hat{s}[n]|^2 = \arg \min_W |x[n] - r[n] \cdot W[n]|^2.$$  

(18)

The initial value of $W$ is an empty matrix $\emptyset$. The update of the NLMS algorithm is described as

$$W[n+1] = W[n] + \alpha \cdot r[n] \cdot \hat{s}[n]/\sigma^2,$$  

(19)

$$\hat{s}[n] = x[n] - r[n] \cdot W[n],$$  

(20)

$$\sigma^2 = r[n] \cdot r'[n]/M + \varepsilon,$$  

(21)

where $\alpha$ is the adaptive factor, $M$ the length of the filter, $\sigma$ the time varying step size, and $\varepsilon$ a small constant to avoid $\sigma$ to be zero.

2.3. Experimental measurements

Seven healthy subjects, 5 males and 2 females with ages varying between 25 and 35 years (mean 29 ± 3.6), volunteered to participate in the experiment. An ECG signal was recorded using two circular Ag/AgCl electrodes of 1-cm diameter (3M RedDot, Nadarzyn, Poland) placed on each forearm of the subject, as shown in Fig. 5. Two accelerometers were placed close to the electrodes to monitor their movements. A Refa amplifier (TMS International, Enschede, The Netherlands) was used to acquire both signals at a sampling frequency of 2048 Hz. An extension cord was placed close to the leads and powered to simulate a poor electrical environment, which may occur in real ECG measurements.

The ECG measurements were performed under two different test conditions: vibrating the arm with different force and manually pulling the electrode lead. The first condition was designed to measure the relationship between MA and PLIM, while the second was designed to evaluate the proposed adaptive filter. To perform the first measurement, a dedicated vibratory setup was employed to generate well controlled vibration to be applied to the arm (Xu et al. 2013). The core of the setup is a motor generating a constant baseline force with superimposed sinusoidal force modulation. A 50-cm aluminum bar is mounted perpendicularly to the motor output shaft to vertically apply the generated force to the arm (Fig. 5). Accurate control of the generated
vibrating force is achieved by dedicated system calibration (Xu et al. 2012). Our preliminary measurements also showed a linear relation \( R = 0.98 \) between the vibrating force and the acceleration measured on the bar. Therefore, in the rest of this paper, the vibration force is used as a measure of the intensity of the movement induced to the arm.

For each measurement, the ECG signals were recorded using two different lead-ground configurations: actively shielded leads + poor grounding (patient ground on the nail of the left thumb) and unshielded leads + good grounding (patient ground on the ankle of the right leg). The first configuration was designed to simulate the model in Fig. 3, assuming the currents coupling into the leads (Fig. 1) are negligible due to shielding. The second configuration was designed to simulate the model in Fig. 1, since the interference caused by the common mode potential (Fig. 3) is negligible due to the high input impedance of the adopted Refa amplifier.

2.3.1. ECG measurements with sinusoidal vibration The subjects were seated comfortably with both arms relaxed to avoid the influence of muscular activity on the recorded signal. The left arm of the subject was well supported while the right arm was positioned on top of the vertically vibrating bar of the adopted vibratory setup (Fig. 5). Five trials of 30 s were performed, where a 7-Hz sinusoidal vibration was applied to the right arm by the vibrating bar. The adopted baseline of the vibration was 0 N, while the vibration amplitude in the trials varied between 20, 30, 40, 50, and 60 N.

2.3.2. ECG measurements with manual pull of the electrode lead All the subjects performed a second 30-s trial without using the vibratory setup. The subjects were seated comfortably with both arms placed on a table. Instead of employing the vibratory setup, electrode movements were generated by manually pulling the lead of one electrode. This measurement simulates movements similar to those in a real ECG recording situation, and was therefore used to test the performance of the proposed adaptive NLMS filter.

2.4. Validation

2.4.1. Spectral analysis In order to analyze the relation between MA and PLIM, spectral analysis was first performed on all the raw ECG signals recorded during sinusoidal vibration. The central ten seconds (10-20 s) of the 30-s signals were used for analysis. A Fast Fourier Transform was applied to each of the 10-s segments. The spectral amplitude at 7 Hz was extracted as an estimation of MA amplitude, while the average amplitude of the two modulated components located at \( f_p \pm 7 \) Hz was extracted as an estimation of PLIM amplitude. For each subject, 5 MA amplitudes and 5 PLIM amplitudes were estimated, as the subject performed 5 trials at different vibrating force. The estimated MA and PLIM amplitudes for each subject were then normalized with respect to their respective maximum value, in order to reduce possible bias caused
by differences among subjects.

2.4.2. Motion artifact removal The effect of the proposed adaptive NLMS filter (Fig. 4) for MA removal was evaluated on the ECG signals recorded while manually pulling the lead. The same interval used for the spectral analysis was adopted for MA removal. The low-pass and band-pass filters mentioned in the block scheme in Fig. 4 were implemented by 4th order Butterworth filters. The PLI removal was implemented by a band-stop \( (f_p \pm 10 \text{ Hz}) \) filter. In order to demodulate \( b[n] \), local maxima of \( b[n] \) were first detected, resulting in a sampling frequency equal to \( f_p \). The detected local maxima were then re-sampled by spline interpolation to obtain the same sampling frequency as the original signal. For the FIR adaptive filter, \( M = 20 \), \( \alpha = 0.001 \), and \( \varepsilon = 0.1 \).

For evaluation, the proposed method was compared with an acceleration-based adaptive filtering method, which is established in the literature (Widrow et al. 1975, Tong et al. 2002, Thakor & Zhu 1991, Xu et al. 2013, Thakor 1987, Luo & Tompkins 1995, Degen & Jackel 2008, Serteyn et al. 2015). The same \( M, \alpha, \) and \( \varepsilon \) values used in the proposed method were employed to implement the acceleration-based adaptive filter. To quantify the difference between the two filtered signals, the mean square error (MSE) between the two signals was calculated and normalized with respect to the variance of the signal obtained using the acceleration-based method, as given by

\[
\text{MSE} = \frac{1}{N} \sum_{n=1}^{N} \left( \hat{s}_1(n) - \hat{s}_2(n) \right)^2 / \sigma_2^2,
\]

where \( \hat{s}_1(n) \) is the clean ECG signal obtained using the proposed method, \( \hat{s}_2(n) \) the signal obtained using the acceleration-based method, \( \sigma_2^2 \) the variance of \( \hat{s}_2(n) \), and \( N \) the number of samples. All the signal processing was implemented in Matlab® 2014b (MathWorks, Natick, MA).

3. Results

3.1. Spectral analysis

The spectral analysis applied to the signals recorded during 7-Hz sinusoidal vibration showed clear components at \( f = 7 \text{ Hz} \) (MA), \( f = f_p \), and \( f = f_p \pm 7 \text{ Hz} \) (PLIM), as shown in Fig. 6. For the signals recorded using shielded leads with poor grounding, in which the effect of current coupling into the body is dominant, both MA and PLIM increased with increasing vibration force up to 50 N, and then slightly decreased at 60 N. The normalized average results over all the subjects are shown in Fig. 7 (a). Furthermore, a high correlation coefficient \( (R = 0.97) \) between MA and PLIM was found \( (p < 0.01) \).

For the signals recorded using unshielded leads with good grounding, in which the effect of current coupling into the leads was dominant, both MA and PLIM were found to increase with increasing vibration force, as shown in Fig. 7 (b). The correlation coefficient between MA and PLIM was 0.98 \( (p < 0.01) \).
3.2. Motion artifact removal

The proposed adaptive filtering scheme (Fig. 4) was applied to the ECG signals recorded during manual pull of the electrode lead. Figure 8 shows a representative example of the signals (shielded leads + poor grounding) in the lower branch of the adaptive scheme, including the raw signal, the signal after band-pass filtering $b_n$, and the demodulated signal $r_n$ after re-sampling. It is clear that the amplitude of the band-pass filtered signal, $b_n$, varies over time, indicating the modulation effect as described in (10).

Figure 9 shows the results after applying the proposed adaptive filtering scheme to the raw ECG signal shown in Fig. 8 (a). The movement-induced baseline wander can clearly be observed in the unfiltered signal (Fig. 9 (a)) and is effectively removed by the proposed method (Fig. 9 (b)). Furthermore, the ECG signal obtained using the
Figure 8. Example of signals in the lower branch in Fig. 4 (shielded leads, poor grounding): a) Raw ECG signal; b) $f_p \pm 10$ Hz band-pass filtered signal; c) Band-pass filtered signal after demodulation.

The proposed adaptive filtering method was comparable to that obtained using acceleration-based adaptive filtering, as shown in Fig. 9 (c). The average MSE between the two clean signals over all subjects (7 recordings) accounted for only $1.0 \pm 1.4\%$ of the variance of the one obtained by acceleration-based adaptive filtering.

Similar results were found for the signals recorded using unshielded leads with good grounding. A representative example is shown in Fig. 10. The average normalized MSE between the two clean signals was $1.4 \pm 2.4\%$.

4. Discussion

This study presents a novel adaptive filtering method for MA reduction in biopotential recordings, by estimating the movement-induced electrode-skin impedance variation as a reference signal. Different from previous adaptive filtering methods that monitor the electrode-skin impedance variation by injecting an external current (Degen & Jackel 2008, Serteyn et al. 2015), we estimate the electrode-skin impedance variation by demodulating the PLI in the proposed method. Therefore, no current injection nor additional sensor is used in the proposed method, in which lies the main novelty of the present study. Although the performance of the proposed method was evaluated using ECG measurements, its application is not limited to the ECG, but can be extended to many other electrophysiological recordings. However, different biopotential recordings may have different properties in both amplitude and frequency and, therefore, the performance of the proposed method may vary for different signals.

By applying a 7-Hz sinusoidal vibration with different amplitude to the arm, our spectral analysis suggested, for both lead-grounding configurations, the 7 Hz and $f_p \pm 7$ Hz components to increase with increasing force, except for a drop at 60 N for the signals recorded with shielded leads and poor grounding. However, high correlation
coefficients between the 7 Hz and $f_p \pm 7$ Hz components were observed for both lead-ground configurations, confirming our first hypothesis that PLIM can reflect movement-induced electrode-skin impedance variation and may be used to extract a reference signal for MA removal.

The drop of the MA and PLIM amplitude at 60 N for the signals recorded with shielded leads and poor grounding cannot be explained by the nonlinear behaviour of the mechanical vibration transmission, as observed in the acceleration signals. In fact, the fundamental components of the acceleration signal were found to be dominant and to increase with increasing vibration amplitude for both lead-ground configurations. Moreover, the ECG signals recorded with unshielded leads and good grounding following the same protocol showed no drop at 60 N. One possible explanation may then relate to the nonlinear behaviour of the amplifier caused by saturation, which is due to the high common mode potential resulting from the high ground impedance. Indeed, clear saturation distortion can be observed in the raw signals recorded with shielded leads and poor grounding, as shown in Fig. 8 (a).

In order to test the performance of the proposed adaptive filter for MA removal, additional trials were performed by manually pulling on the lead in order to simulate MA similar to that occurring in real measurements. The proposed adaptive filter effectively removed MA for both lead-ground configurations, confirming the reliability of
the proposed method for MA reduction. Furthermore, the performance of the proposed adaptive MA removal scheme is comparable to the acceleration-based adaptive filtering method, further confirming the accuracy of the proposed method. However, for both methods, a distortion of the T wave may be observed after filtering (Fig. 10), which may be due to an overlap between the frequency band of the T wave and that of movement, detected either by the accelerometer or by demodulation of the PLMI. As a result, any signal in the frequency band of movement may be removed by adaptive filtering, which is a common problem with MA removal by adaptive filtering. However, the focus of the present study is mainly on avoiding the use of additional sensors in adaptive filtering rather than solving the distortion problem.

The presence of the PLI is essential for MA removal in the proposed method. For measurements with perfect grounding and active shielding, most PLI is rejected, which may limit the application of the proposed method. However, in many situations, such as continuous ambulatory monitoring, optimal use of adequate grounding and/or active shielding is limited due to their requirement of hardware reduction and low power consumption. As a result, the PLI is present in the final recordings, enabling the possibility to apply the proposed method for MA removal.

Indeed, the proposed method is a passive solution for MA removal in continuous ambulatory monitoring. One reason is that the proposed adaptive MA removal method extracts the reference signal from the PLIM without additional sensors, further meeting the requirement of hardware reduction and low power consumption. Furthermore, in ambulatory monitoring, MA may be much more critical as compared to the PLI. Therefore, even in case the PLI is not sufficiently strong to extract the reference signal for MA removal, the PLI can be intentionally increased and then removed after extracting the reference signal.

In the present study, we focus mainly on the adaptive MA removal, by extracting a reference signal from the PLI. Although it is not the objective of this study, the PLI and PLIM also need to be removed after obtaining the reference signal. A band-stop filter was therefore implemented in the present study to remove the PLI and PLIM from the signal of interest. A stop band \(f_p \pm 10\) Hz was adopted by taking into consideration the maximum frequency of body movement (10 Hz) (Tam & Webster 1977, Burbank & Webster 1978). However, in many applications, body movement occurs at relatively lower frequency and, therefore, a narrower stop band may be employed. In particular, in situations where PLIM are not dominant, the band-stop filter can be replaced by a notch filter with center frequency at \(f_p\), which removes PLI only, minimizing the suppression of the signal of interest. Furthermore, the PLI and PLIM can also be removed by using a dedicated adaptive scheme, as described in (Martens et al. 2006).

Finally, it should be noted that our measurements were performed with stable PLI. In case biopotentials are recorded in a fastly changing electrical environment, the PLI may show amplitude variations. As a result, the reference signal extracted by demodulating the PLI may contain components reflecting not only movement-induced electrode-skin impedance variation but also PLI amplitude variation. However, possible
variations of PLI amplitude, represented by variations in the amplitude of the reference signal, are correlated neither to the desired ECG signal nor to MA and, therefore, should not affect the performance of the proposed adaptive method for MA removal. On the other hand, it should be noted that PLI amplitude variations may affect the convergence property of the adaptive filter, as the adaptive filter needs to adjust its coefficients in response to variations in the amplitude of the reference signal.

5. Conclusion

In the present study, a novel adaptive filtering method is proposed for MA reduction in biopotential recordings based on the power-line interference. Based on our results we may conclude that the spectral components around \( f_p \) are highly correlated with MA and can be used to extract a reference signal for the adaptive filter for MA removal. Evaluation of the proposed method on ECG signals showed effective MA removal, comparable to the acceleration-based adaptive filtering method. This method brings new insight in MA removal, enabling reliable analysis of biopotential recordings in ambulatory settings.

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