COMPARISON BETWEEN BLIND SEPARATION AND ADAPTIVE NOISE CANCELLATION TECHNIQUES FOR FETAL ELECTROCARDIOGRAM EXTRACTION

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Abstract. This contribution addresses the problem of fetal electrocardiogram (FECG) extraction from maternal cutaneous potential measurements. One blind source separation method based on higher-order statistics and Widrow's multi-reference adaptive noise cancelling method are applied to real multi-channel ECG recordings obtained from a pregnant woman. The results demonstrate the more robust performance of the blind technique in this biomedical application.

1 Introduction

Monitoring the fetus' heart condition during pregnancy in order to diagnose possible diseases is crucial. An early diagnosis before delivery increases the effectiveness of any subsequent treatment, which calls for non-invasive techniques. Among them, the fetal electrocardiogram (FECG) is acknowledged by the medical community as a reliable diagnosis tool. The *antepartum* FECG is usually obtained through skin electrodes attached to the mother's body. The physician observes at the electrode output the desired fetal heartbeat contributions buried in an additive mixture of undesired disturbances: mains coupling, thermal noise from electronic equipment, mother's respiration and electromyographic (EMG) signals (e.g., owing to an uncomfortable position of the patient, uterus contraction, etc.) and, most important, the maternal ECG (MECG) components, of considerably higher amplitude than the fetal signals. This mixture of interference hinders the observation of the FECG, and hence the diagnosis procedure. Appropriate signal processing techniques are thus required in order to recover the wanted FECG components from the corrupted potential recordings.

A variety of different approaches have been proposed to address this problem. Classical procedures such as coherent averaging, matched filtering, auto- and cross-correlation based methods, adaptive filtering, sequenced adaptive filters, etc. have not proven totally satisfactory [2, 10]. It is shown in [8] that this biomedical application can be reframed in a more efficient manner from the perspective of *blind source separation (BSS)*, thus implying that BSS techniques may be applied to tackle the FECG extraction problem. The BSS methods described in [2] and [10] rely on the second-order statistics (SOS) of the data. This type of procedures, generically known as *principal component analysis (PCA)* [14], show a separation quality strongly dependent on a careful electrode placement selection. This obstacle is surmounted by exploiting the higher-order statistics (HOS) and looking for higher-order independence at the separator output [8, 9, 13], yielding a general class of methods known as *independent component analysis (ICA)* [14]. See [4] for the original mathematical definition of the ICA.

Results obtained from the application of ICA-BSS methods to the biomedical problem in hand have already been reported [1, 3, 8, 9, 13, 14], as well as contrasted with other PCA-BSS methods [1, 8, 13, 14]. The primary objective of the present contribution is to establish a comparison between a specific HOS-based BSS method and one of the most significant classical techniques proposed to solve this challenging problem: Widrow's multi-reference adaptive noise cancelling (MRANC) method [11].

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2 Experimental Data

Fig. 1 displays the real recordings used in the experiments.[1] The signals were recorded over 10 seconds at a sampling frequency of 500 Hz, yielding a total of $T = 5000$ samples per signal, from 8 cutaneous electrodes located on different points of a pregnant woman's body. Specifically, the first five recordings correspond to electrodes on the mother's abdominal region. In them a mixture of FECG, MECG and noise is visible. The last three signals were digitized from the mother's thoracic region, and no FECG heartbeat component can be perceived, due to the longer distance between these electrodes and the fetal heart. Remark that only the relative, rather than the absolute, amplitudes of the vertical axes are relevant.

3 Methods

3.1 Multi-Reference Adaptive Noise Cancellation

The adaptive filtering solution proposed by Widrow and colleagues in their famous 1975 paper [11] was one of the the first successful approaches to the FECG extraction problem. An abdominal lead $a(k)$ acts as a primary input to an adaptive noise canceller (Fig. 2), where k denotes a discrete-time index. This input mainly contains a mixture of FECG and MECG signals. The MECG interference that corrupts the abdominal lead is considered as the 'noise' to be eliminated. The reference inputs to the canceller are n thoracic leads, mostly composed of maternal heartbeat components, and are represented by the N-sample vectors $\mathbf{r}_k^{(i)} \triangleq [r^{(i)}(k),\, r^{(i)}(k-1),\, \ldots,\, r^{(i)}(k-N+1)]^\dagger$, $i = 1, \ldots, n$, symbol † standing for the transpose operator. These inputs are processed by means of n finite impulse response (FIR) filters of tap-length N, characterized by time-varying coefficient vectors $\mathbf{w}_k^{(i)} \triangleq [w_0^{(i)}(k),\, w_1^{(i)}(k),\, \ldots,\, w_N^{(i)}(k)]^\dagger$, $i = 1, \ldots, n$. The output at instant k is then given by $\varepsilon(k) = a(k) - s(k)$, with $s(k) = \sum_{i=1}^n \mathbf{w}_k^{(i)\dagger} \mathbf{r}_k^{(i)} = \sum_{i=1}^n \mathbf{r}_k^{(i)\dagger} \mathbf{w}_k^{(i)}$. Aiming to reduce the MECG disturbance at the canceller output, the filter coefficients are updated so as to minimize the output power, $\mathrm{E}[\varepsilon^2]$. An implementation more suitable for real-time processing is attained via the steepest descent method over the instantaneous stochastic approximation of the true gradient. This yields the well-known least mean squares (LMS) update [5, 6, 11], which in our multi-reference set-up reads:

$$\mathbf{w}_{k+1}^{(i)} = \mathbf{w}_k^{(i)} + 2\mu_k^{(i)} \varepsilon(k) \mathbf{r}_k^{(i)}, \qquad i = 1, \ldots, n. \tag{1}$$

Parameter $\mu_k^{(i)}$ is the adaption coefficient or step-size of the ith filter at time k. To improve the algorithm robustness with respect to the input power, the updating term in (1) can be divided by $N\hat{P}_k^{(i)}$, where $\hat{P}_k^{(i)}$ is an estimate of the ith reference-signal power calculated as:

$$\hat{P}_{k+1}^{(i)} = (1 - \tilde{\mu}_k^{(i)})\hat{P}_k^{(i)} + \tilde{\mu}_k^{(i)} r^{(i)}(k)^2. \tag{2}$$

This variant, which is shown to provide smaller misadjustment than the original method at a faster convergence rate and rather similar computational cost, is a particular version of the so-called power-normalized LMS (PNLMS) algorithm [5].

3.2 Blind Source Separation

By virtue of an electrical model of the heart [2, 8, 9, 13], each of the p electrodes located on the patient's body, say $\mathbf{y}(k) = [y_1(k),\, y_2(k),\, \ldots,\, y_p(k)]^\dagger \in \mathbb{R}^p$, can be considered as an instantaneous linear combination of the different sources of bioelectric current. All bioelectric activity is modelled by means of q unobservable independent *source signals*, $\mathbf{x}(k) = [x_1(k),\, x_2(k),\, \ldots,\, x_q(k)]^\dagger \in \mathbb{R}^q$. Then, the electrode measurement vector accepts the matrix relationship:

$$\mathbf{y}(k) = M\mathbf{x}(k). \tag{3}$$

[1]The authors are grateful to L. De Lathauwer, D. Callaerts and J. Vandewalle, from K. U. Leuven, Belgium, for providing the recordings.

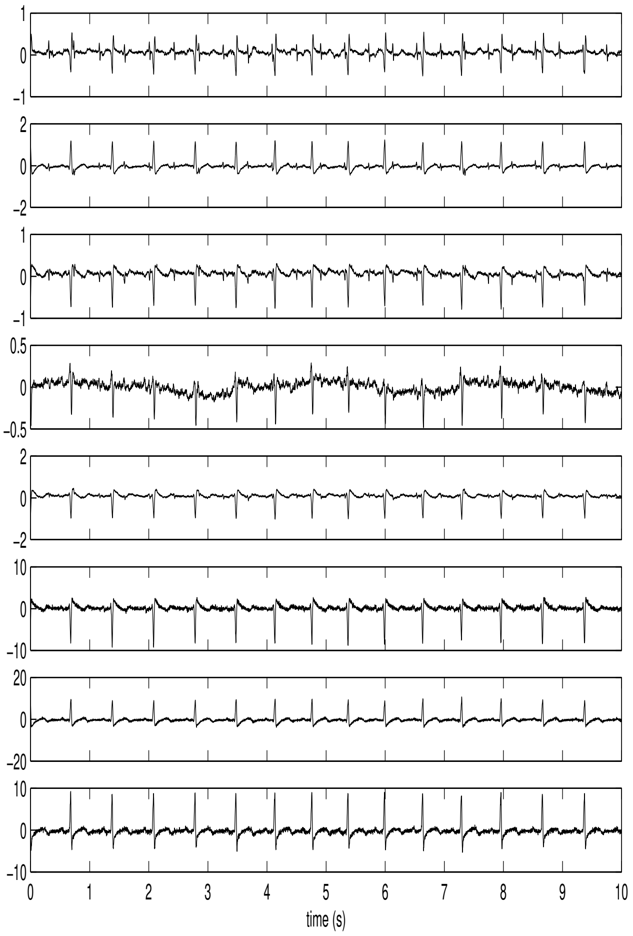

Figure 1: A cutaneous electrode recording from a pregnant woman.

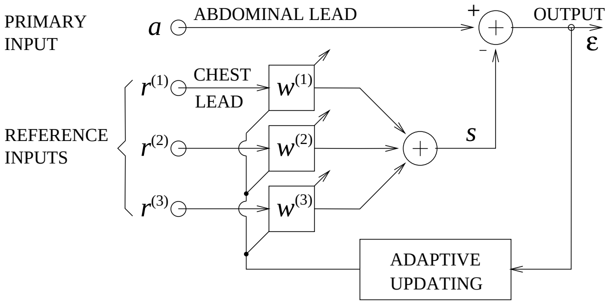

Figure 2: MRANC solution to the FECG extraction problem.

Matrix $M \in \mathbb{R}^{p\times q}$ is called *mixing matrix*, and its structure is determined by the body geometry, the electrode and source positions and the conductivity of the body tissues [10]. Equation (3) corresponds to the familiar BSS model of instantaneous linear mixtures [14].

Assuming that the surface potentials are generated according to model (3), the reconstruction of the FECG contributions to the recordings is reduced to the identification of the independent sources of fetal cardiac activity and the corresponding elements of the mixing matrix. This goal requires the estimation of the source signals $\mathbf{x}$ and the mixing matrix M. In these experiments, we consider the BSS method developed in [15] (see also [16] for its adaptive implementation). Essentially, it is a two-step procedure, consisting of an initial PCA stage (also known as *pre-whitening*) followed by a higher-order processing part aiming at higher-order independence. The latter stage is composed of elementary Givens transformations applied in turn to each pre-whitened signal pair $[z_i,\, z_j]^\dagger$. Rotation angles $\hat{\theta}$ characterizing such transformations are computed in closed form as:

$$\hat{\theta} = \tfrac{1}{4}\,\mathrm{angle}\big(\xi\cdot\mathrm{sign}(\gamma)\big), \quad \text{with} \left\{ \begin{array}{l} \xi = \mathrm{E}[\rho^4 \mathrm{e}^{\jmath 4\phi}] \\ \gamma = \mathrm{E}[\rho^4] - 8, \end{array} \right. \tag{4}$$

where $\rho\,\mathrm{e}^{\jmath\phi} = z_i + \jmath\, z_j$ and $\jmath^2 = -1$. Symbol $\mathrm{E}[\cdot]$ denotes the mathematical expectation, or ensemble average, which in practice is replaced by time averages over the signal samples. This pairwise process is repeated over the $p(p-1)/2$ signal pairs until convergence, usually taking about $1 + \lfloor\sqrt{p}\rfloor$ sweeps (that coincides with the value originally found for the method of [4]). Note that this procedure is indeed based on HOS, since the expectations in (4) can be expressed as a function of the data fourth-order cumulants.

4 Experimental Methodology and Results

For Widrow's MRANC solution, the three thoracic leads of Fig. 1 (last 3 recordings) are employed as reference inputs to the canceller (i.e., $n = 3$, as depicted in Fig. 2), whereas the abdominal leads play (in turn) the role of primary inputs. The number of filter taps, for simplicity and due to the identical nature of the reference signals, is chosen to be equal for all filters. Standard guidelines given in [12] yield $N = 60$ taps. This provides a total delay, with uniformly spaced taps, of 120 ms.[2] The other parameters required by the PNLMS algorithm described in Section 3.1 are selected as: $\mu_k^{(i)} = \mu = 10^{-2}$ and $\tilde{\mu}_k^{(i)} = \tilde{\mu} = 2 \cdot 10^{-3}$, $\forall i$, $\forall k$.[3] The resulting waveforms obtained from each of the five abdominal leads are shown in Fig. 3.

On the other hand, Fig. 4 shows the FECG contributions to the abdominal electrodes obtained from the recordings of Fig. 1 by the BSS method. Two FECG sources were found in the set of estimated sources, and isolated in vectors $\hat{\mathbf{x}}_f = [\hat{x}_{f_1}, \hat{x}_{f_2}]^\dagger$. Matrix $\hat{M}_f = [\hat{\mathbf{m}}_{f_1}, \hat{\mathbf{m}}_{f_2}]$ contained the corresponding columns of the estimated mixing matrix. The fetal heartbeat contributions to the recordings were then calculated through:

$$\hat{\mathbf{y}}_f = \hat{M}_f \hat{\mathbf{x}}_f. \tag{5}$$

Observe that this procedure allows the estimation of FECG contributions to *all* leads, even the thoracic ones. Nevertheless, only the abdominal signals are shown here for the sake of a more meaningful comparison with the MRANC.

[2] cf. 129 ms of the experiments reported in [11, Section VIII-C], where log-periodic tap spacing is employed.

[3] The choice of $\tilde{\mu}$ ensures convergence of (2) to the true power value around $k = T/2$ for stationary signals.

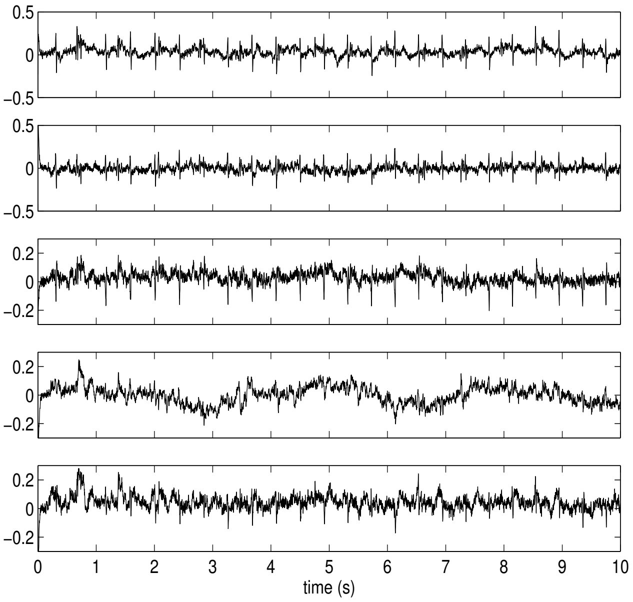

Figure 3: Fetal contributions to the abdominal electrodes (first five signals) of Fig. 1 obtained by the MRANC method.

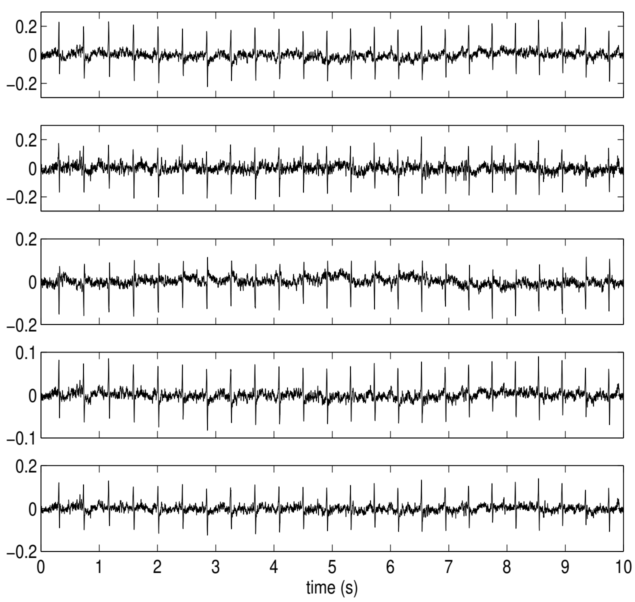

Figure 4: FECG contributions to the abdominal electrodes (first five signals) of Fig. 1 obtained by the BSS method.

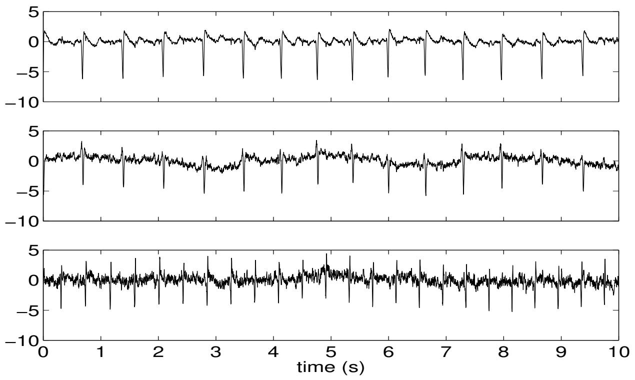

Figure 5: Source signals recovered by the BSS method from only 3 abdominal electrodes (waveforms 3, 4 and 5) of the ECG recordings of Fig. 1. Three sweeps over the signal pairs were performed.

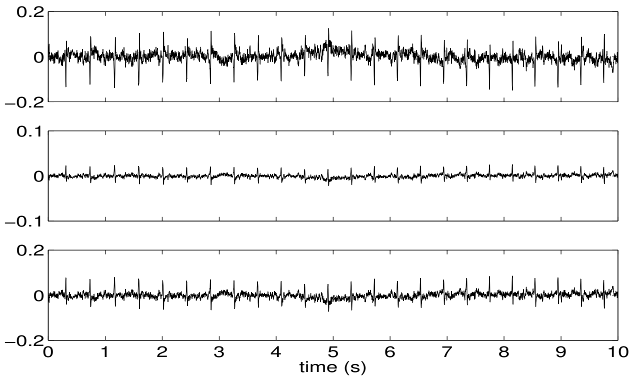

Figure 6: FECG contributions to electrodes 3, 4 and 5 obtained from the source signals of Fig. 5.

5 Discussion

Out of the waveforms recovered by the MRANC method in Fig. 3, only the 1st and 3rd signals display fairly clear fetal heart components. MECG interference still corrupts the 2nd output, while the 4th signal is heavily corrupted by noise and a periodic component, presumably due to the mother's respiration. Since there is hardly any of this periodic component present in the chest leads, it cannot be filtered and subtracted from the 4th abdominal electrode, where it appears. The results from this approach, therefore, seem to be very dependent on the primary electrode placement, similarly as occurs with SOS-based BSS techniques [2, 10].

The quality of all the FECG contributions reconstructed by the higher-order BSS method is notably superior. For instance, the periodical component which corrupted the 4th signal obtained by the MRANC solution (Fig. 3) is now completely eliminated in Fig. 4, since it is found to be generated by an independent source of interference and hence it is extracted from the 4th recording as a separate noise signal. The elimination of the perturbations caused by the MECG can be explained likewise.

The BSS robustness with respect to the electrode placement is evidenced through further experiments whereby FECG sources are clearly revealed by exploiting abdominal leads exclusively. Good results were obtained by processing the output of up to only 3 electrodes, as illustrated in Figs. 5 and 6. In Fig. 5, the fetal and maternal subspaces are successfully separated, and a FECG source (3rd signal) practically free from MECG interference is recovered. From the fetal source, the FECG contributions to the electrodes involved are obtained as explained in Section 4, and are displayed in Fig. 6 (cf. last 3 plots of Figs. 3 and 4). The dimension of the estimated fetal subspace is now lower than in the full-recording processing, which accounts for the dissimilar results in both cases. However, the FECG extraction achieved from such smaller number of electrodes can still be considered as satisfactory.

Results obtained in this biomedical context by other BSS methods are reported in [1,3,8,10,13,14,17]. In [1], [13] and [14] the recordings of Fig. 1 are processed through the PCA and the techniques of [4] (HOEVD) and [7] (HOSVD). Basically, results from the BSS method evaluated here are much more precise than from the PCA and the HOSVD, and very similar to the HOEVD. In [17] the experiments of this paper are completed with another dataset, obtaining analogous outcomes.

In conclusion, HOS-based BSS proves a more robust, successful approach to the non-invasive FECG extraction problem than the MRANC, although the superior performance is attained at the expense of an increased computational complexity. Nonetheless, the achieved FECG-extraction quality offers promising prospects for the use of ICA-BSS techniques in prenatal medical diagnosis.

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