On-line Fetal Heart Rate Monitoring Using SQUID Sensor Arrays

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ABSTRACT

Objective: Fetal magnetoencephalography (fMEG) is the only complete non-invasive method to record fetal brain signals in the utero. When using this method in a clinical setting, it is necessary to monitor the fetal heart rate during the recordings. Based on the high sensitivity of the magnetic sensors used, it is not possible to record the heart rate simultaneously with fMEG by conventional methods like Doppler ultrasound. However, based on a large sensor array, the maternal and fetal magnetocardiographic signals (mMCG and fMCG) are recorded simultaneously with the fMEG with adequate temporal and spatial resolution. Method: Recordings were performed on a 151 channel SQUID array covering the whole maternal abdomen (SARA system). We use a multichannel adaptive filter to cancel the maternal signal and dynamically select those channels with high content of fetal signal. Our technique can be implemented in real time using a recursive steepest descent algorithm. From the extracted fMCG signal, we obtain an on-line estimate of the heart rate with an R-wave detector that uses a smoothed first derivative magnitude of the signal. Results: We show through real data that, with this technique, we obtain real-time estimates of the fetal heart rate with a small standard deviation. Furthermore, our proposed criterion for channel selection can identify the channels with low or high content of fetal signal. Conclusion: We developed a fast and reliable fMCG extraction method, which has on-line capability. Currently, initial studies are conducted with an experimental setup, allowing the on-line access to the magnetic signals of all 151 channels. Our method will help to establish fetal magnetometry in a clinical setting, where continuous control of vital signals of the fetus is necessary.

KEYWORDS

Fetal magnetoencephalography, magnetocardiography, adaptive filtering, signal cancellation.

INTRODUCTION

Fetal magnetoenchepalography (fMEG) is a completely passive and non-invasive technique for the study of fetal brain function *in utero* [Lenge, 2001]. fMEG is measured in the presence of environmental noise and various near-field biological signals: maternal magnetocardiogram (mMCG), fetal magnetocardiogram (fMCG) uterine smooth muscle (magnetomyogram), and motion artifacts [Wakai, 2002]. In clinical studies, it is necessary to monitor the fetal heart rate during the recordings. However, traditional techniques such as fetal electrocardiogram (fECG), uterine pressure sensors, or Doppler ultrasound, do not provide with complete information of the cardiac function.

fMCG has multiple advantages for the analysis of the morphology and temporal parameters of the fetal heart signals [Lewis, 2003]. It fills the diagnostic gap left by the difficulties involved in recording fECG due to the insulating effect of the *vernix caseosa* and the existence of preferred conduction pathways between the fetal heart and maternal abdomen. Furthermore, fMCG is detected as early as the 16th week of gestation, and measurements can be obtained simultaneously with fMEG using a sufficiently large sensor array and adequate sample rate.

However, the detection of the fMCG signal depends on the ability to separate the fetal heart component from the composite signal. In general, the mMCG and fMCG signals are the dominant components of the ensemble, with an average magnitude as large as 10 pT at the fetal thorax location. Closer to the maternal heart, the mMCG can be as large as 100pT [Vrba, 2003]. Therefore, algorithms for heart rate monitoring based on MCG have to overcome not only this problem, but need to have a low computational complexity and low processing delay, and must be able to detect rapid and dramatic changes in the heartbeats in a beat-to-beat manner.

In this paper, we present an algorithm based in a multichannel filter to adaptively separate the fMCG signal from the mMCG, as well as dynamically select those channels with highest content of fetal signal and monitor the fetal heart activity on-line.

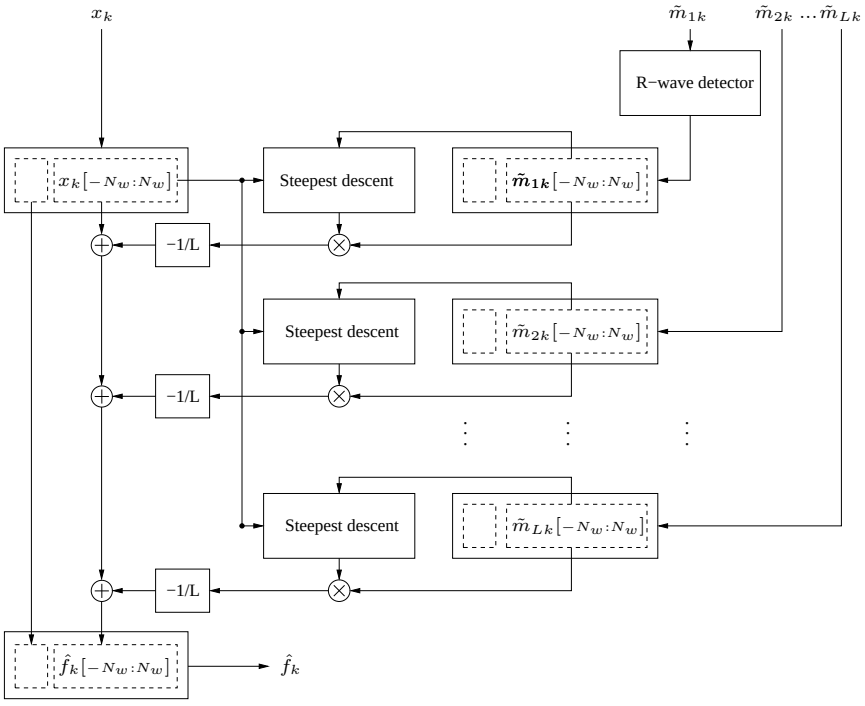

Figure 1: Diagram of the multichannel adaptive filter.

METHODS

We define our measurement model as $x(n) = f(n)+m(n)$, where $x(n)$ is the measurement of a single channel, $f(n)$ is the fetal signal component, $m(n)$ is the maternal signal component plus noise, and $n = 1, 2, \ldots, N$ time samples. Our goal is then to cancel $m(n)$ from $x(n)$. This problem can

be seen from the point of view of adaptive signal processing as a case of *signal cancellation* [Hayes, 1996] in which the process $f(n)$ is to be estimated from a corrupted observation $x(n)$.

Consider that a group of L channels can be used as *reference*, i.e. they contain mostly maternal signal. Refer to each of the reference measurements as $\tilde{m}_i(n)$, where $i = 1, 2, \ldots, L$. Although $\tilde{m}_i(n)$ will be correlated with $m(n)$, the two processes will not be equal. Therefore, it is not possible to estimate $f(n)$ simply by subtracting $\tilde{m}_i(n)$ from $x(n)$. Instead, we use an adaptive filter to estimate $m(n)$ and then subtract it from $x(n)$, i.e. $\hat{f}(n) = x(n) - \hat{m}(n)$, where $\hat{f}(n)$ and $\hat{m}(n)$ are the estimates of the fetal and maternal signals, respectively, at the nth time sample. If we assume that we can temporarily store N_T time samples of our on-line measurements in a buffer, then we can define $\boldsymbol{x}_k = [x(n_0), x(n_0+1), x(n_0+2), \ldots, x(n_0+N_T-1)]^T$, and $\tilde{\boldsymbol{m}}_{ik} = [\tilde{m}_i(n_0), \tilde{m}_i(n_0+1), \tilde{m}_i(n_0+2), \ldots, \tilde{m}_i(n_0+N_T-1)]^T$, where n_0 is an arbitrary starting time sample such that $n_0 + N_T - 1 \leq N$ at the kth buffer stored. Now, the estimate of the maternal signal at the kth buffer is given by $\hat{\boldsymbol{m}}_k = (1/L)\sum_i w_{ik}\tilde{\boldsymbol{m}}_{ik}$, where w_{ik} is the adaptive coefficient corresponding to the ith reference signal. We use the *steepest descent* algorithm to compute the adaptive coefficients recursively while keeping the mean square error of $\hat{\boldsymbol{f}}_k$ minimized. According to this algorithm, our adaptive coefficients are given by $w_{i,k+1} = (1+\mu\hat{\sigma}^2_{\tilde{m}})w_{ik} + \mu\hat{\sigma}^2_{x\tilde{m}}$, where μ is the step size, $\hat{\sigma}^2_{\tilde{m}}$ is the sample variance of $\tilde{\boldsymbol{m}}_{ik}$, and $\hat{\sigma}^2_{x\tilde{m}}$ is the sample covariance between $\boldsymbol{x}_k$ and $\tilde{\boldsymbol{m}}_{ik}$.

Accurate on-line removal of the maternal signal requires all calculations to be done in synchrony with the peak produced by the maternal heart beat. The detection of this peak is achieved through an R-wave detector that uses a smoothed first derivative of the signal [Köhler, 2002], i.e. $\Theta = 0.4\max\{[0.25, 0.5, 0.25]^T * |\tilde{\boldsymbol{m}}'_{1k}|\}$, where Θ is the threshold of the detector, and $\tilde{\boldsymbol{m}}'_{1k}$ is the first derivative of $\tilde{\boldsymbol{m}}_{1k}$. Note that only the first of the reference channels is passed through the R-wave detector as we consider that the delay between the reference signals is negligible. Once we have detected the peak of the maternal signal, all computations are done for windows of $2N_w + 1$ time samples from both $\boldsymbol{x}_k$ and $\tilde{\boldsymbol{m}}_{ik}$: N_w samples before and N_w after the peak, where N_w is the window size.

Finally, the estimate of the fetal signal at the kth iteration is obtained as $\hat{\boldsymbol{f}}_k = \boldsymbol{x}_k - \hat{\boldsymbol{m}}_k$. A complete diagram of our adaptive filter is shown in Fig. 1. Further details of our method can be found at http:\\cyclone.ece.uic.edu\~david\onlinefHRM.

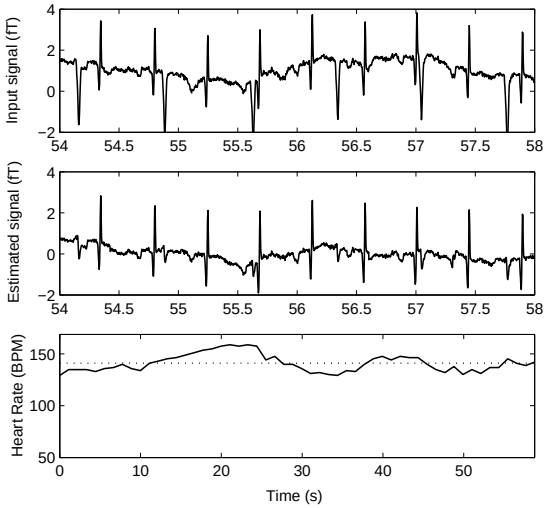

Figure 2: From top to bottom: Input signal ($\boldsymbol{x}_k$), output signal ($\hat{\boldsymbol{f}}_k$), and estimate of the fetal heart rate (mean shown in dotted line).

EXAMPLE

We applied our adaptive filter to real magnetic data from a 151 channel SQUID array covering the whole maternal abdomen (CTF SARA system). Our data was sampled at 312.5 Hz and then low-pass filtered at a cut-off frequency of 40 Hz. The 7 upper rim channels were used as our reference channels $\tilde{m}_i(n)$, i.e. $L = 7$. We considered a buffer size $N_T = 300$ and a window size $N_w = 80$. These values were chosen accordingly to the Nyquist rate. For the steepest descent algorithm, we chose a step size $\mu = 0.1$.

Our preliminary results showed that $|w_{ik}| \to \mu$ for those channels with high content of fetal signal. Using this criterion, we chose the signal from one of these channels to estimate the heart rate. This estimate was obtained at each window through an R-wave to detect the peaks of the estimated fetal signal $\hat{\boldsymbol{f}}_k$. Such R-wave detector was implemented in the same way as explained in the previous section. A snapshot of $\boldsymbol{x}_k$ and $\hat{\boldsymbol{f}}_k$ for the selected channel, as well as the estimated fetal heart rate, are shown in Fig. 2.

CONCLUDING REMARKS

We have presented a reliable and fast method with on-line capabilities to extract the fMCG signal. Our method is based in an adaptive filter to cancel the maternal signal component. This technique also allowed us to identify those channels with poor, regular, or high content of fetal signal by keeping track of the values of the adaptive coefficients. This capability may be also used to track on-line changes of the fetus position and allow adaptive channel selection. Finally, using this procedure for a specific channel, we can obtain an accurate estimate of the fetal heart rate. Our method will help to establish fetal magnetometry in a clinical setting, where continuous control of vital signals of the fetus is necessary.

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