Maternal ECG removal from non-invasive fetal ECG recordings

Rik Vullings, Chris Peters, Massimo Mischi, Guid Oei, and Jan Bergmans

Abstract— Fetal monitoring during pregnancy is important to support medical decision making. The fetal electrocardiogram (fECG) is a valuable signal to diagnose fetal well-being. Non-invasive recording of the fECG is performed by positioning electrodes on the maternal abdomen. The signal to noise ratio of these recordings is relatively low and the main undesired signal is the maternal electrocardiogram (mECG). Existing methods to remove the mECG signal are not sufficiently accurate to extract the complete fECG signal. In this paper, a novel method for removal of the mECG signal from abdominal recordings is presented. It is an extension of the linear prediction method. Each mECG complex is segmented and these segments are separately estimated by linear prediction. Both the presented method and the standard linear prediction are applied to simulated abdominal recordings and evaluated by determining the rms errors between the estimated and the actual fECG signals. The ratio between the rms errors of the linear prediction method and the presented method varies between 0.4 dB and 2.3 dB. It can therefore be concluded that the presented method is capable of a more accurate removal of the mECG signal for all simulated abdominal recordings with respect to the linear prediction method.

I. INTRODUCTION

A major problem in modern obstetrics with respect to fetal monitoring is the lack of possibilities to extract information from the fetus to assess its condition. The fetal heart rate is one of the few useful signals that can be measured non-invasively. In clinical practice it is often the only source of information that is available. During pregnancy, the fetal heart rate is monitored using Doppler ultrasound. Unfortunately, this technique is inaccurate and provides a relatively low positive predictive value [1]: it is reliable only when the condition of the fetus is clearly good or clearly bad. In different situations, additional tests, such as microblood examinations, are required. Furthermore, extensive use of Doppler ultrasound for fetal heart rate monitoring might be harmful to the fetus.

During labor, it is possible to monitor the fetal heart rate by means of an electrode directly connected to the fetal scalp. This scalp electrode provides a more reliable fetal heart rate and provides additional information by means of the fetal electrocardiogram (fECG). Changes in the condition of the

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fetus are reflected in the ECG as changes in the morphology. Therefore, determination of the fECG in addition to the fetal heart rate can improve prediction of fetal distress. Drawback of the direct measurement of the fECG is that it is an invasive technique that can only be applied during labor when the fetal membranes have ruptured.

In stages of pregnancy earlier than labor, it could be possible to monitor the fECG by means of electrodes positioned on the abdomen of the mother. Furthermore, by using multiple electrodes it could be possible to reconstruct the fetal vectorcardiogram. This vectorcardiogram, in fact, is a valuable diagnostic tool in electrocardiography [2]. Because of the low signal to noise ratio of these measurements, additional signal processing techniques are required to extract the fECG.

The signals measured at the abdomen are a mixture of electrophysiological signals and noise. Examples of electrophysiological signals present in the recordings are the maternal electrocardiogram (mECG), the electrohysterogram (EHG, the electrical activity of the uterus) and the fECG. The spectrum of the EHG typically ranges between 0.1 Hz and 3 Hz [3]. Therefore most of the EHG can be suppressed by applying a linear phase highpass filter with a cut-off frequency at 2 Hz; filtering with a higher cut-off frequency would affect the fECG. One of the main sources of noise is the power line signal, which can be suppressed by applying a notch filter.

As the amplitude of the fECG is small with respect to the amplitude of the mECG and the spectra of both ECG signals overlap [4], the mECG cannot be suppressed by the application of a classical filter. Several techniques to extract the fECG from abdominal recordings have been proposed. Commonly, methods based on independent component analysis (ICA) and linear prediction (LP) are used [5][6]. ICA determines the linear transformation that maximizes the statistical independency of the transformed signals [7]. Due to the low signal to noise ratio of the fECG, in most situations only one or two [8] of the transformed signals represent the fECG signal. The fact that only one of two transformed signals represent the fECG signal is a significant drawback that limits the diagnostic value of the method, since it does not allow the accurate reconstruction of the fetal vectorcardiogram, as for this reconstruction several fECG signals are required [9]. Therefore, ICA is not further considered in our approach to obtain the fECG.

The estimation of the fECG by the LP method (and other averaging methods) is based on the subtraction of the average mECG signal. This is an advantage with respect to ICA as the linear prediction method extracts the fECG signal

for each electrode, enabling the reconstruction of the fetal vectorcardiogram. Due to variability in the shape of the ECG, the averaged mECG complexes are not completely equal to the actual recorded mECG complexes. Main causes for this variability are the respiratory effects [2]. As a result, the fECG signals obtained by subtraction of the averaged mECG complexes from the recorded signals are affected by residuals of the mECG. These residuals can have amplitudes of the same magnitude as the amplitude of the fECG and can lead to errors.

In this paper we present a new method to extract the fECG signals from abdominal recordings. This method is based on the subtraction of an averaged mECG. In contrast to the LP method this method averages each wave of the ECG complex separately, instead of averaging the ECG complex as a whole. Therefore, the developed method is referred to as segmentational linear prediction (SLP). The SLP method and the LP method are evaluated by applying both methods on simulated abdominal recordings. These simulated abdominal recordings are composed of actual recordings, but in contrast to standard abdominal recordings the fetal signal is known as it is measured directly using a scalp electrode. After subtraction of the mECG it is therefore possible to evaluate both algorithms. The fECG signals obtained by the SLP method show an improved agreement with the actual fetal signals with respect to the fECG signals obtained by the LP method.

II. METHODOLOGY

A. Measurement

The fECG measurements are performed by positioning 12 Ag/AgCl electrodes in two horizontal lines on the maternal abdomen and one electrode on each of the maternal shoulders (Fig. 1). The recorded signals are digitized using a Porti-16/ASD amplifier (TMS International B.V., the Netherlands) operating at 400 Hz sampling rate with 22 bit resolution. The fECG is measured by positioning an electrode on the fetal scalp. The signal from this electrode is recorded using a HP 8040A (Agilent, Palo Alto, California) second-generation fetal CTG-monitor and digitized by a NI 6034E data acquisition board operating at 1024 Hz sampling rate with 16 bit resolution.

To evaluate the performance of the SLP method with respect to the performance of the LP method, nine abdominal recordings are simulated. The bipolar lead of the mECG measured at the shoulders, resembling Einthoven I, is used as maternal signal and the fECG measured by the scalp electrode is used as fetal signal. Prior to generating the simulated signals, the fECG signal is downsampled to 400 Hz. The first four simulated abdominal recordings are characterized by the rms amplitude ratio between the fECG signal and the mECG signal. These ratios are 0, -3, -10, and -13 dB. The next four simulated abdominal recording have a rms amplitude ratio of -10 dB between the fECG signal and the mECG signal and are superimposed on a simulated Gaussian noise signal, representing instrumentation (and muscle) noise. The rms amplitude ratios between the noise and the mECG are 0, -3,

-10, and -13 dB. The ninth simulated abdominal recording is obtained from another patient, having a larger variability in the maternal heart rate with respect to the other simulated recordings (79 $\pm$ 8 beats per minute and 71 $\pm$ 3 beats per minute, respectively) and has a fECG/mECG rms amplitude ratio of -10 dB. Fig. 2 shows a part of the third simulated abdominal recording.
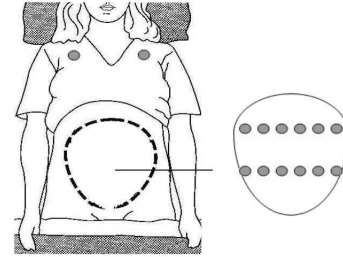

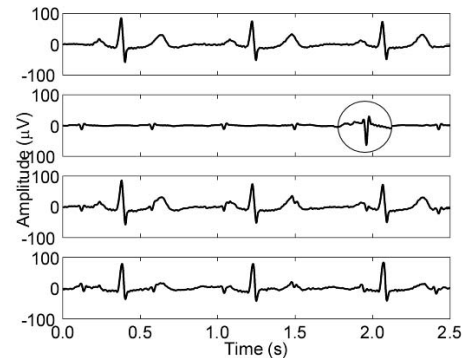Fig. 1. Positions of the electrodes on the maternal abdomen and maternal shoulders.

Fig. 2. Simulated abdominal recording. The upper window shows the maternal signal, the second window shows the fetal signal. In the circle an enlargement of the fetal signal is displayed. The third window shows the simulated recording, while the bottom window shows an abdominal recording that is measured simultaneously.

B. Preprocessing

The power line signal is suppressed by the application of a fourth order Butterworth notch filter [10]. The EHG is suppressed by applying a fourth order Butterworth linear phase high-pass filter with a cut-off frequency at 2 Hz.

C. Removal of the maternal ECG

Fig. 3 shows the block diagram of both the LP method and the SLP method.

The simulated recording $\vec{x}$ is composed of a mECG signal, a fECG signal, and noise. Individual mECG complexes $\mathbf{X}_{n,m}$ are determined by detecting the R-peaks (Fig. 4) in the simulated recording and subdividing $\vec{x}$ in n mECG complexes of m samples. Detection of the R-peaks is performed by means of a peak detection algorithm that is based on the length transformation [11]. Both the LP method and the SLP method work by averaging N successive mECG complexes to estimate the shape of the next mECG complex $\mathbf{X}_{i,1...m}$, i.e. the ith row of matrix $\mathbf{X}$.
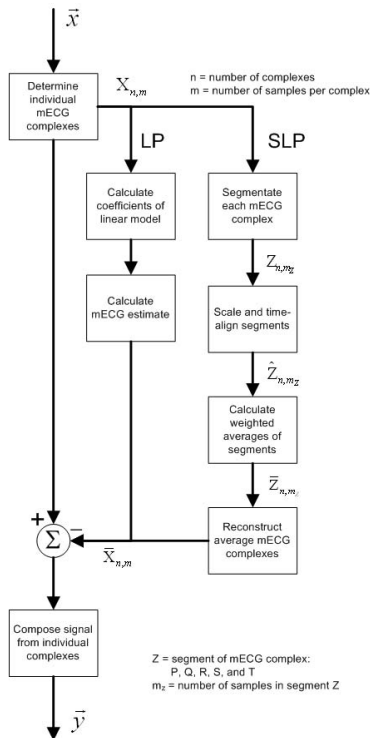
Fig. 3. Block diagram of the LP method and the SLP method.

1) The linear prediction method: The LP method determines the coefficients a_i of the linear model by minimizing the squared error ε, which is given as:

$$\varepsilon = \sum_{j=1}^{m}(\mathbf{X}_{i,j} - \sum_{k=1}^{N}(a_{i-k}\mathbf{X}_{i-k,j}))^2. \quad (1)$$

The average mECG complex $\overline{\mathbf{X}}_{i,1...m}$ is calculated as:

$$\overline{\mathbf{X}}_{i,1...m} = \sum_{k=1}^{N}(a_{i-k}\mathbf{X}_{i-k,1...m}). \quad (2)$$

2) the segmentational linear prediction method: Depending on the position of the electrode with respect to the maternal heart, the ECG complex is composed of the combination of a P-wave, representing the depolarization of the atria, a QRS-complex representing the depolarization of the ventricles, and a T-wave representing the repolarization of the ventricles. These waves are shown in Fig. 4. At first, the algorithm determines which of these waves are present. This determination is performed by detecting the start and end of each wave by means of adaptive thresholds. The start and end of each wave are indicated in Fig. 4 as transitions between solid and dashed lines. If the width and amplitude of the detected wave satisfy a physiological model, the wave defined as being present.

After determination of the individual waves $\mathbf{Z}_{n,m_z}$ ($\mathbf{Z} = \mathbf{P}, \mathbf{Q}, \mathbf{R}, \mathbf{S}, \mathbf{T}$), average waves $\overline{\mathbf{Z}}_{n,m_z}$ are calculated for each mECG complex. Averaging is performed in two steps. In the first step the waves of N preceding mECG complexes are time-aligned and scaled to minimize the mean squared error with respect to the corresponding wave $\mathbf{Z}_{i,1...m_z}$ of the
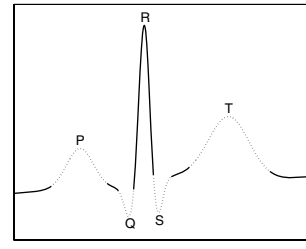

Fig. 4. The five waves that are present in the mECG. Each transition between solid and dashed lines indicates the start or end of a wave.

current mECG complex. The time-aligned and scaled waves $\hat{\mathbf{Z}}_{i-k,m_z}$ are given as:

$$\hat{\mathbf{Z}}_{i-k,j} = A\mathbf{Z}_{i-k,j+B} + C, \quad (3)$$

with $j = 1...m_z$, i the index for the current wave, and $1 \leq k \leq N$. The scaling factor A, the horizontal time-alignment B and the vertical offset C are calculated by minimizing the mean squared error between each of the N preceding waves and the current wave:

$$\left(\frac{\partial}{\partial A}, \frac{\partial}{\partial B}, \frac{\partial}{\partial C}\right) \sum_{j=1}^{m_z}((\mathbf{Z}_{i,j} - \hat{\mathbf{Z}}_{i-k,j})^2) = (0,0,0), \quad (4)$$

for each $1 \leq k \leq N$. To solve this equation, both the preceding waves and the current wave are parabolically interpolated to generate a continuous function.

In the second step, an average wave is calculated as the weighted linear combination of the time-aligned and scaled waves, where the weights w_N are given by the reciprocals of the mean squared errors between the preceding waves and the current wave:

$$w_k = \left(\sum_{j=1}^{m_z}(\mathbf{Z}_{i,j} - \hat{\mathbf{Z}}_{i-k,j})^2\right)^{-1}. \quad (5)$$

Using linear regression for the calculation of the weights is considered as an alternative. However, the weights determined by linear regression can be negative and as a result the shape of the average wave can adapt to the shape of the wave that is estimated. When the wave that is estimated comprises a superposition of a mECG wave and a fECG complex, the average mECG complex contains part of the fECG. The fECG will therefore be affected by subtraction of the average mECG.

Each average wave $\overline{\mathbf{Z}}$ is thus composed of a weighted linear combination of the waves of N preceding mECG complexes:

$$\overline{\mathbf{Z}}_{i,j} = \frac{\sum_{k=1}^{N} w_k \hat{\mathbf{Z}}_{i-k,j}}{\sum_{k=1}^{N} w_k}, \quad (6)$$

with $j = 1...m_z$. The average mECG complexes $\overline{\mathbf{X}}$ are reconstructed by combining the separate average waves.

By subtracting the average mECG complexes $\overline{\mathbf{X}}$ from the mECG complexes $\mathbf{X}$ and combining the individual complexes to rebuild the simulated recording, the mECG is

removed from the simulated abdominal recording $\vec{x}$, leaving the signal $\vec{y}$ that is composed of the fECG signal and noise.

III. RESULTS

Both the SLP method and the LP method are applied on simulated abdominal recordings of 100 seconds. For one simulated abdominal recording the fECG signals that are obtained by both methods are partly shown in Fig. 5. Fig. 5 also shows the fECG signal that is used in the simulated abdominal recording.
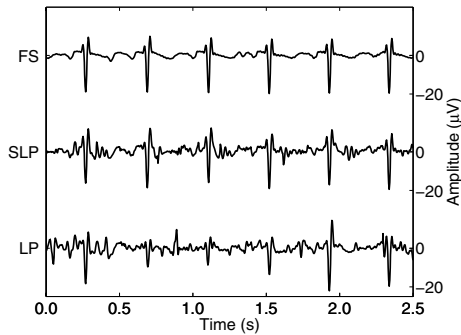
Fig. 5. FECG signals obtained by both methods for the simulated abdominal recording with fECG/mECG amplitude ratio of -10 dB. The upper signal is the fetal signal (FS) used for the simulated abdominal recording. The center signal is the fECG obtained by the SLP method and the bottom signal is the fECG obtained by the LP method.

Both methods are evaluated by calculating the rms error between the resulting fECG signals and the actual fECG signals. For all simulated recordings, the amplitude of the fECG signal is kept constant. Fig. 6 shows the calculated rms errors for the SLP method and the LP method. Furthermore, Fig. 6 shows the ratio between these rms errors.
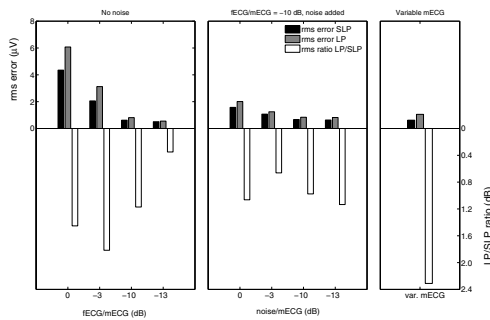
Fig. 6. Calculated rms errors between the fECG signals obtained by the SLP and LP methods and the actual fECG signals. The first four simulated abdominal recordings have amplitude ratios between the fECG and mECG signals of 0, -3, -10, and -13 dB. The next four simulated abdominal recordings have amplitude ratios of -10 dB and are superimposed on a Gaussian noise signal. The amplitude ratios between the noise and the mECG are 0, -3, -10, -13 dB. In the ninth simulated abdominal recording a mECG signal from another patient is used, having a larger variability in the maternal heart rate with respect to the heart rate variability in the other simulated recordings (79 ± 8 beats per minute and 71 ± 3 beats per minute, respectively).

IV. DISCUSSION AND CONCLUSIONS

The rms errors between the fECG signals obtained with the SLP and LP methods and the actual fECG signals are smaller for the SLP method in all simulated abdominal recordings. The ratio of the rms errors for both methods varies between 0.4 dB and 2.3 dB. Based on this result, it can be concluded that the SLP method is capable of removing the mECG signals more accurately than the LP method for all the simulated abdominal recordings. From figure 6 it can be concluded that the optimal fECG/mECG ratio for subtraction of the mECG is approximately -10 dB, which is a common amplitude ratio for non-invasive fECG recordings. For larger ratios the fECG is not completely suppressed in the average mECG waves and is therefore affected by subtraction of the average mECG. For smaller ratios, the rms errors between the average mECG and the actual mECG have the same order of magnitude as the fECG.

As expected, the addition of noise results in an improvement of the performance of the LP method with respect to the SLP method. Reason for this is that the addition of noise causes the determination of the individual mECG waves to be less accurately. The ratio between the rms errors for the simulated abdominal recording with the larger variability in the maternal heart rate is increased with respect to the rms errors ratio for the third simulated abdominal recording, which has the same fECG/mECG amplitude ratio. Reason for this is that the SLP method is more capable of handling variability in the mECG signal than the LP method, because each wave of the mECG signal is averaged separately. Future plans include the calculation of bipolar fECG complexes to study the possibility of deriving the fetal vectorcardiogram.

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