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Heart Rate Detection in Low Amplitude Non-Invasive Fetal ECG Recordings

Chris Peters, Rik Vullings, Jan Bergmans, Guid Oei, Pieter Wijn

Abstract—Multi-electrode electrical measurements on the maternal abdomen may provide a valuable alternative to standard fetal monitoring. Removal of the maternal ECG from these recordings by means of subtracting a weighted linear combination of segments from preceding maternal ECG complexes, results in fetal ECG traces from which the fetal heart rate can be determined. Unfortunately, these traces often contain too much noise to determine the heart rate by R-peak detection. To overcome this limitation, an algorithm has been developed that calculates the heart rate based on cross-correlation. To validate the algorithm, noise was added to a fetal scalp ECG recording to simulate low amplitude abdominal recordings. Heart rates calculated by the algorithm were compared to the heart rates from the original scalp ECG. For simulated signals with a signal to noise ratio of 2, the coefficient of correlation was 0.99 ($p < 0.001$). By using the developed algorithm for calculating the fetal heart rate, multi-electrode electrical measurements on the maternal abdomen now can be used for fetal monitoring in relatively early stages of pregnancy or other situations where ECG amplitudes are low or noise levels are high.

I. INTRODUCTION

THE fetal heart rate is an essential parameter in monitoring the condition of the unborn child. It is one of few sources of information that are available and is practically the only parameter that can be used for online monitoring. In standard fetal monitoring, the fetal heart rate is measured by either Doppler ultrasound measurements or by ECG recordings using an electrode that is placed directly on the fetal scalp. Doppler ultrasound measurements are non-invasive, but often provide inaccurate results [1]. Scalp electrode ECG recordings provide a reliable means of monitoring the fetal heart rate, but can only be applied during labor, when the fetal membranes have ruptured. An additional drawback of the directly measured fetal ECG is that it is an invasive measurement, which causes discomfort to the mother and may cause injury to the fetus.

A more reliable alternative to Doppler ultrasound measurements and a more patient-friendly and safer

alternative to scalp ECG measurements is provided by multi-electrode electrical measurements on the maternal abdomen. From these measurements, fetal ECG traces can be retrieved after removing the maternal ECG by subtracting a weighted linear combination of separate waves of preceding ECG complexes [2]. In relatively late stages of pregnancy, including labor, the fetal heart rate can be determined from a linear combination of resulting fetal traces by standard R-peak detection. In earlier stages of pregnancy, when the fetal heart is much smaller, the amplitude of the fetal QRS complex is much lower than at the end of pregnancy ($5\mu\text{V}$ and below). Therefore, in these stages of pregnancy it is often not possible to determine the fetal heart rate by R-peak detection. Additional factors like a malnourished fetus, an obese mother, or substantial noise can also make R-peak detection impossible. To overcome these limitations of non-invasive fetal ECG measurements, an algorithm has been developed that calculates the fetal heart rate from these recordings in an alternative way. The algorithm is based on cross-correlating fetal traces obtained after subtracting the maternal ECG from abdominal recordings. To validate the algorithm, artificial signals with low signal to noise ratios are constructed, containing a previously recorded scalp ECG signal. The heart rates calculated by the algorithm are compared to the heart rates determined by R-peak detection on the original scalp ECG signal, which is considered as golden standard.

II. METHODOLOGY

A. Algorithm

Four traces resulting from maternal ECG subtraction are used to calculate the fetal heart rate. First, each of the traces is filtered by a 4th order Chebyshev bandpass filter [3] that operates between 15 and 30 Hz. This way, most noise outside the frequency content of the fetal QRS complex is suppressed [4]. Next, each trace is processed in segments of fixed length. For each trace, a specific segment is cross-correlated with the corresponding segments of the other traces. Then the envelopes of these cross-correlations are calculated and then summed. By detecting the peaks in the summed envelopes, the time-shift that corresponds to the time between consecutive heart beats can be determined, similar to the calculation of the fetal heart rate from Doppler ultrasound measurements [5].

In this study, signals were repeatedly analyzed while varying the segment length from 1.5 to 5 s. This way, insight is

C.H.L. Peters is with the Department of Clinical Physics, Máxima Medical Center, Veldhoven, The Netherlands; e-mail: chl.peters@mmc.nl.

R. Vullings is with the Faculty of Electrical Engineering, Eindhoven University of Technology, Eindhoven, The Netherlands.

J.W.M. Bergmans is with the Faculty of Electrical Engineering, Eindhoven University of Technology, Eindhoven, The Netherlands.

S.G. Oei is with the Department of Gynaecology and Obstetrics, Máxima Medical Center, Veldhoven, The Netherlands.

P.F.F. Wijn is with the Department of Clinical Physics, Máxima Medical Center, Veldhoven, The Netherlands and the Faculty of Applied Physics, Eindhoven University of Technology, Eindhoven, The Netherlands.

provided in which segment length is optimal. During signal analysis, the segment window is shifted in steps of 0.25 s. The final processing step is the application of a moving average filter with window length equal to the segment length. Physiologically impossible heart rates are excluded during averaging.

B. Signal generation

To simulate noisy abdominal fetal ECG traces, a fetal ECG trace from a scalp recording was used. This trace contained 1215 ECG complexes and showed healthy variations in heart rate. Noise that is similar to noise in actual abdominal fetal ECG measurements was added to the scalp signal. This noise was generated by applying the filters used in the processing of the abdominal signals [2] to Gaussian white noise. Signals were constructed with signal to noise ratios of 1, 2, and 4. Signal to noise is defined as:

$$SNR = \frac{P_{signal}}{P_{noise}} = \frac{V_{rms,signal}^2}{V_{rms,noise}^2} \quad (1)$$

For ECG recordings, it is not representative to use the power in the complete signal. Instead, an estimate of the average power in the QRS complexes in the recording is used. For this estimate, the peak-to-peak amplitude of all QRS complexes in the signal is determined. Assuming a sinusoid shape of the QRS complex, $V_{rms,signal}$ becomes:

$$V_{rms,signal} \approx \frac{1}{2} \sqrt{2} \cdot \frac{1}{2} \cdot V_{pp,QRS} \quad (2)$$

This results in a signal to noise ratio of:

$$SNR = \frac{\frac{1}{8} \cdot V_{pp,QRS}^2}{V_{rms,noise}^2} \quad (3)$$

where $V_{pp, QRS}$ is the mean peak-to-peak amplitude of the QRS complexes in the scalp signal and $V_{rms,noise}$ is the rms amplitude of the noise that was added to this signal.

Fig. 1 shows three seconds of a fetal ECG trace that is obtained by removing the maternal ECG signal from a bipolar measurement on the maternal abdomen. In this trace, the fetal ECG has low amplitude but is still partly visible.

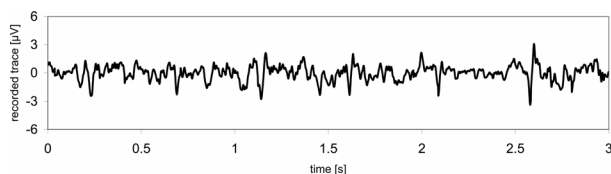


Fig. 1. Non-invasively measured fetal ECG trace with low amplitude. The fetal heart rate in this trace is approximately 120 BPM.

Fig. 2a shows three seconds of the scalp ECG trace that was used for signal generation. Fig. 2b shows three seconds of

generated noise. Fig. 2c shows the sum of the signals in Fig. 2a and Fig. 2 b, the signal to noise ratio of this signal, calculated using (3), is 2.

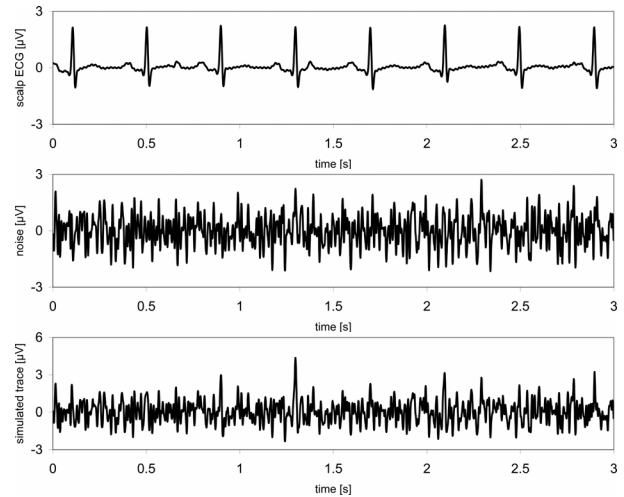


Fig. 2a. Scalp ECG recording, scaled to low amplitude. (upper graph)
 Fig. 2b. Filtered Gaussian white noise. (middle graph)
 Fig. 2c. Simulated non-invasive fetal ECG trace, created by summing the signals in Fig. 2a. and Fig. 2b. (lower graph)
 The signal to noise ratio of this simulated trace equals 2. Simulated signals were also generated with SNR = 1 and SNR = 4.

C. Calculated heart rate validation

The heart rates calculated by the algorithm can be validated by investigating the correlation with the heart rates determined from the original scalp ECG trace. To enable direct comparison of the heart rates of both methods, fetal heart rates are determined from the scalp ECG trace by R-peak detection and subsequently resampled at a frequency of 4 Hz [6]. This way, fetal heart rates from the scalp ECG trace and simulated signals are available at identical time bases.

For each of the three sets of signals (SNR=1, SNR=2, and SNR=4) heart rates are calculated for eight different segment lengths (1.5, 2, 2.5, 3, 3.5, 4, 4.5, and 5s). The coefficients of correlation between these results and the fetal heart rate from the scalp ECG are calculated. Also, the percentage of time that the algorithm was not able to determine a fetal heart rate is calculated.

III. RESULTS

Fig. 3a shows the heart rate that is determined from the original scalp ECG measurement by R-peak detection. Fig. 3b shows the heart rate calculated by the cross-correlation algorithm for the signals with SNR=2 and a segment length of 2.5 s.

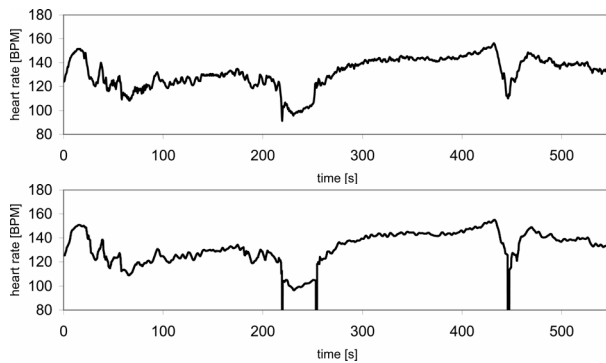


Fig. 3a. Fetal heart rate determined by R-peak detection on the original scalp ECG.

Fig. 3b. Fetal heart rate calculated by the developed algorithm. Simulated signals with a SNR of 2 were used. A segment length of 2.5 s was used.

Fig. 4 shows the heart rate calculated from the simulated signals plotted against the heart rate determined from the scalp ECG in a scatter plot.

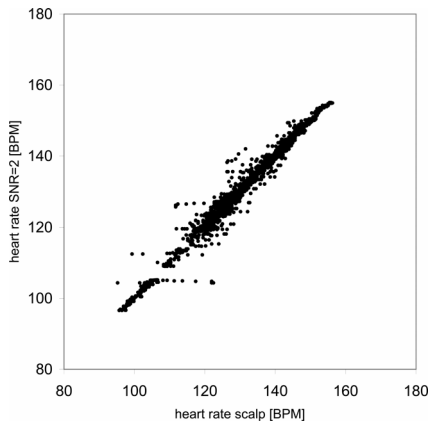


Fig. 4. Scatter plot of heart rates determined by the cross-correlation algorithm vs. the golden standard: the scalp ECG. Note the gap between 105 and 110 BPM on the y-axis; The actual heart rate however is in this range three times very shortly ($t=220$ s, $t=255$ s, and $t=450$ s). All three times the algorithm is not able to determine the heart rate.

For all analyzed datasets, the coefficients of correlation R and the percentage of time the algorithm was not able to determine a heart rate are presented in Table I.

TABLE I
COEFFICIENTS OF CORRELATION R AND FAILURE PERCENTAGES

Segment length (s)	SNR = 1		SNR = 2		SNR = 4	
	R	% fail	R	% fail	R	% fail
1.5	-	19.4	0.88	2.13	0.99	0.32
2	0.75	5.53	0.99	1.63	0.99	0.45
2.5	0.95	4.85	0.99	0.54	0.99	0.27
3	0.96	1.63	0.99	0.54	0.99	0.27
3.5	0.94	2.13	0.98	0.50	0.94	2.13
4	0.89	2.86	0.98	0	0.99	0.55
4.5	0.87	2.50	0.98	0.64	0.98	0.41
5	0.87	2.32	0.97	2	0.96	1.59

For SNR=1 and a segment length of 1.5 s, no coefficient of correlation could be calculated. For all other datasets the p-value was below 0.001.

IV. DISCUSSION AND CONCLUSIONS

For all three sets of signals, best results are achieved for segment lengths between 2.5 and 3.5 s. If the segment length is short, the segment contains only a small number of heart beats and therefore the calculation is relatively sensitive to noise. This is confirmed by the improvement in results for short segment lengths for higher signal to noise ratios. If the segment length is much longer, the segment contains a large number of heart beats. As the heart rate of a healthy fetus is fluctuating continuously, the performance of the algorithm gets worse for longer segments. This is illustrated in Fig. 3a and Fig. 3b, where the fetal heart rate changes quickly at $t=220$ s, $t=255$ s and $t=450$ s and the algorithm is not able to provide an accurate heart rate. Also, the high frequency variation that can be seen in the heart rate determined from the scalp ECG trace is less in the heart rate calculated by the algorithm. Segment lengths between 2.5 and 3.5 s typically contain five to ten heart beats, and appear to be optimal for the performance of the algorithm.

The results show that a signal to noise ratio of 1 is too low for accurate calculations of the fetal heart rate. However, the algorithm is capable of calculating the heart rate in signals with a signal to noise ratio of 2. For signal to noise ratios of 4, the results are even better, but in these signals R-peak detection will probably provide similar results.

The developed algorithm provides a valuable means to calculate the heart rate from signals where standard methods fail. This makes it possible to use multi-electrode non-invasive electrical measurements on the maternal abdomen for fetal monitoring in relatively early stages of pregnancy and other situations where ECG amplitudes are low or noise levels are high.

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