Estimating the Shape of the Fetal Pulse Curve for Transabdominal Pulse Oximetry using Synchronous Averaging*

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***Abstract*— A sufficient oxygen supply of the fetus is necessary for a proper development of the organs. Transabdominal fetal pulse oximetry is a method that allows to measure the oxygenation of the fetal blood non-invasively by placing the light sources and photodetectors on the belly of the pregnant woman. The shape of the measured fetal pulse wave is needed to extract parameters for the estimation of the oxygen saturation. This work presents an extension of our previously presented signal processing strategy that allows to extract an average shape of the fetal pulse wave from noisy mixed photoplethysmograms (PPG) with dominating maternal and very weak fetal signal components. An adaptive noise canceller and a comb filter are used to suppress the maternal component. The quality of the resulting fetal signal is sufficient to identify single pulse waves in time domain. Further processing demonstrates the extraction of the mean shape of a single fetal pulse wave by synchronous averaging of several detected pulses. The method is evaluated with different datasets of several simulated and synthetic signals measured with a tissue mimicking phantom. The feasibility of the approach is demonstrated by preparing the mixed PPGs to perform fetal pulse oximetry in future studies. However, clinical measurements are needed to finally evaluate the proposed system beyond synthetic datasets.**

I. INTRODUCTION

An adequate oxygen supply of the fetus is necessary for its healthy development. The oxygenation of the fetal blood is seen to be an important vital parameter. In clinical practice, a blood sample is taken from the unborn child after rupture of the amniotic sac and analyzed in the laboratory. The values are used to conclude about the wellbeing of the child and the need of a Caesarean section.

Transabdominal fetal pulsoximetry is a non-invasive method to continuously observe the fetal oxygen level [1], [4], [6]. The method is applicable with an intact amniotic sac during the latter half of pregnancy and may be used for long-term monitoring. Light sources and detectors are placed on the belly of the pregnant woman to measure the photoplethysmogram (PPG) as shown is several previous works [4], [5], [2], [3], [6]. The measured signal consists of a mixture of the maternal and fetal pulse curves [4], [5], [10], [6]. One major challenge of the method is the extraction of the fetal pulse curve [5], [10], [8]. Following the principle of pulse oximetry, the resulting fetal PPG needs to be of high quality to clearly identify the signal level during systole and diastole of the fetus to conclude about the oxygen saturation.

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Our previous work focused on the estimation of the fetal heart rate in time and frequency domain by using adaptive noise canceller and comb filters [8], [7]. We were able to extract the synthetically generated fetal signals and to identify single pulses in time domain. In this work we demonstrate an extended signal processing strategy using improved peak detectors and an synchronous averaging procedure to finally get an average shape of fetal pulse curve.

II. METHODS

A. Signal processing strategy

The proposed processing strategy is shown in Fig. 1. The approach is based on the system presented in our previous work [7]. A reference signal of the mother and the abdominally measured mixed signal are fed into an adaptive noise canceller (ANC). The ANC is based on an recursive least squares (RLS) algorithm, using a filter order of 50 and an initial input covariance estimate of 0.00001 [7]. The resulting signal is used to get an estimate of the fetal heart rate in the frequency domain. The power of the maternal signal is reduced, so that it is possible to detect the fetal signal by using a peak detector. To overcome the influence of noise and other disturbances, a multichannel dataset may be used to clearly identify the fetal component. Based on the strategy presented in [7], a comb filter is designed and applied to the signal in order to finally suppress all undesired components except the fetal PPG. It has been demonstrated earlier that the quality of the resulting fetal PPG is sufficient for time domain analysis and single pulse detection [7]. The following processing module added to the current strategy is the synchronous averager. Fig. 2 shows its functional principle. A threshold based peak detector is used to identify the single fetal pulses in the processed signal. The minima prior to the identified maxima are also detected as starting point of the pulse wave. Further, the found positions of the minima and maxima are optimized by matching them between the signals in case of a multi sensor measurement. Detected points that are found in only one of the signals are rejected, applying a threshold of ± 25 ms. Several sections of the signal are extracted by cutting it at these minima. The length of the sections vary because of the pulse rate variability and vague positions of the detected points. A fixed length of the single pulse wave is used to interpolate each section and to avoid this problem. In the last step the extracted signal sections are averaged to get the mean shape of the fetal pulse wave.

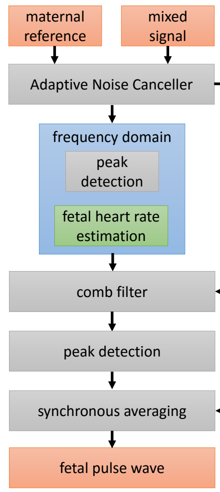

Fig. 1. An adaptive noise canceller, a comb filter and an averager are the main components of the signal processing strategy shown in the flowchart. The adaptive noise canceller suppresses the maternal component in the mixed signal. The fetal heart rate, estimated in frequency domain by peak detection, is used to design a comb filter. A peak detector is used to generate trigger pulses in time domain. The signal is cut by means of these trigger pulses and synchronously averaged. The resulting signal is an average fetal pulse curve.

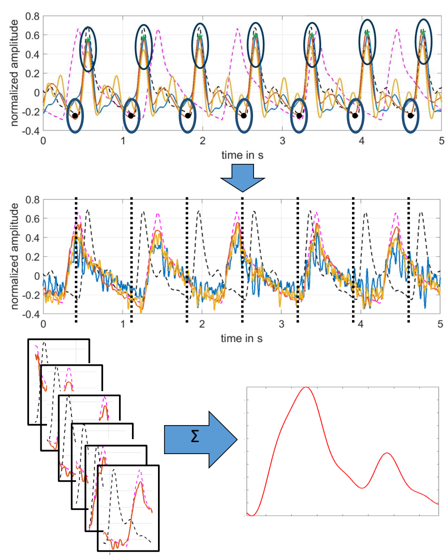

Fig. 2. The principle of the synchronous averaging is shown in this figure. The upper plot shows the comb filtered signals (red, blue and yellow) with the detected fetal pulses (upper ellipses mark the peaks, lower ellipses with points mark the base point) in the time domain. The detected points are used as a trigger to cut the mixed signal in sections as illustrated by the broken lines in the middle plot. The signal sections are added up to get a mean pulse curve as shown in the chart on the bottom.

B. Synthetic datasets

Different kinds of signals are used in this work to demonstrate the performance of the proposed method. A simulated signal consists of one channel and is used to demonstrate the influence of pulse rate variability on the resulting average. The generation of this signal is based on resampling and assembling of single pulse curves, following the explanation in our former work [7]. Three more datasets are measured at the phantom presented by Laqua et al. in [11]. The measurement setups are shown in Fig 3. The configuration of the LED and photosensors is a complex system that is discussed in our previous work [9]. One LED ($\Lambda = 905\ nm$) and three photosensors (Texas Instruments OPT101) are used in Setup 1 and 2. Setup 1 has the LED and the detectors placed in a line symmetrically on the dome of the phantom, realising different source-detector distances. Setup 2 has the LED placed on the tip of the dome and the detectors around it. Each of these two setups provide datasets with three PPG signals. Setup 3 has two LEDs ($\Lambda_1 = 635\ nm$, $\Lambda_2 = 945\ nm$) and one photosensor (Advanced Photonix Inc. PDB-C613-2) in a line, similar to Setup 1. The resulting dataset consists of 2 PPG signals after demultiplexing of the pulsed LED signals. The phantom provides a pressure signal of the maternal and fetal tubes that are used as references for the signal analysis.

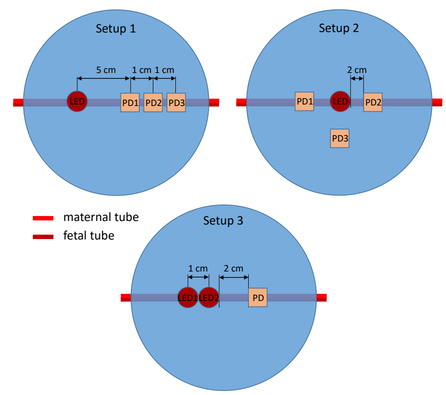

Fig. 3. The figure shows three measurement setups used to generate synthetic signals by the tissue mimicking phantom. The blue circles represent the plan views of the semi-sphere representing the belly of the pregnant woman. The light source (LED) and three photodetectors (PD1, PD2 and PD3) are placed in a line with varying source-detector distances on top of the phantom in Setup 1. Setup 2 has the three detectors placed on a circle around the LED. Setup 3 has two LEDs and one detector in a line, using different wavelengths and source-detector distances.

Sections of the generated signals are shown in Fig 4. The upper two plots show the simulated (artificial) signal, generated with mean maternal and fetal pulse rates of 80 bpm and 137.5 bpm, respectively. The heart rates vary in ranges between 65 bpm to 95 bpm for the mother and 115 bpm to 160 bpm for the fetus. The other three datasets, generated by the phantom, have a maternal pulse rate of 60 bpm in Setup 1 and 45 bpm in Setup 2 and 3 and a fetal pulse rate of 85.7 bpm in all three setups. We chose the lower maternal pulse rate to have a smaller distance between the first harmonic of the maternal signal and the fundamental frequency of the fetus.

The maternal pulse curve dominates in the time domain. In dependence on the measurement setup, the signals are more or less noisy. The artificial signals is superimposed by white Gaussian noise with an SNR of 15 dB. The signals in Setup 1 have a clear maternal pulse curve in time domain. In Setups 2 and 3 the signals are weaker, having a much lower SNR. It is quite difficult to identify the maternal PPG in signals of Setup 3. The components of the signal can be analyzed better in the frequency domain (left handed plots in Fig. 4). The fundamental frequency of the maternal and the fetal component is clearly visible in most signals of the phantom measurements. Signals generated by the phantom do not have a pulse rate variability, so that the maternal and fetal components are represented by small peaks in the frequency spectra. It can be seen that the signals in Setup 1 are disturbed by cross terms, represented by small peaks between the fetal and maternal component. They are analyzed more in detail in our former works in [10], [8]. Due to the heart rate variability, the maternal and fetal component are not separable in the overall frequency spectrum of the artificial signal (upper left plot of Fig. 4). The power of the fetal PPG is spread all over the range of the HRV, so that a notably weaker fetal peak is expected.

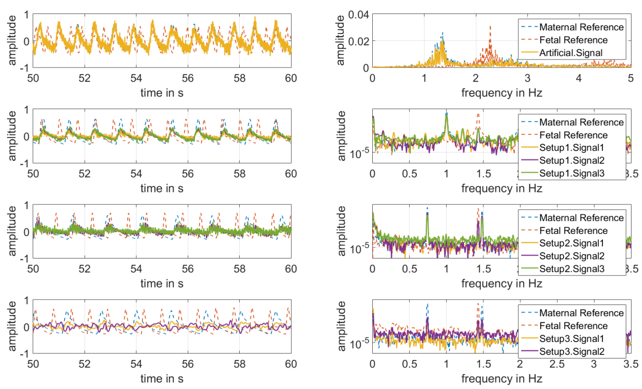

Fig. 4. The charts show the generated signals (left) and their frequency spectrum (right), starting with the artificial signal on top, followed by Setup 1, Setup 2 and Setup 3. The maternal and the fetal reference signal are shown to identify the signal components.

III. RESULTS

The results after using the ANC and the comb filter are shown in Fig 5. The frequency spectrum of the artificial signal after ANC filtering allows to identify the weak fetal component (upper left plot in Fig 5). The peak detector estimates a mean fetal pulse rate of 2.28 Hz (136.8 bpm), that matches the reference signal with an error of 0.01 Hz. After applications of the comb filter, a disturbed fetal pulse wave is visible in time domain. The noisy and disturbed signal leads to a vague peak detection, showing obvious inaccuracies.

The lower plots in Fig 5 show the results of the processed phantom signals. A quite accurate estimate of the fetal pulse rate (1.43 Hz or 85.8 bpm) is feasible because of the sharp peaks in the frequency spectra after ANC filtering. The precise detection in the signals of Setup 1 is well supported by using all three signals for plausibility check, according to the assumption that the fetal pulse rate is the same in all three signals, while the other disturbing sharp peaks close to the fetal and maternal components may vary. One exception is Setup2.Signal3, where the fetal signal is not detectable at all. This may be due to an inconvenient source-detector-configuration.

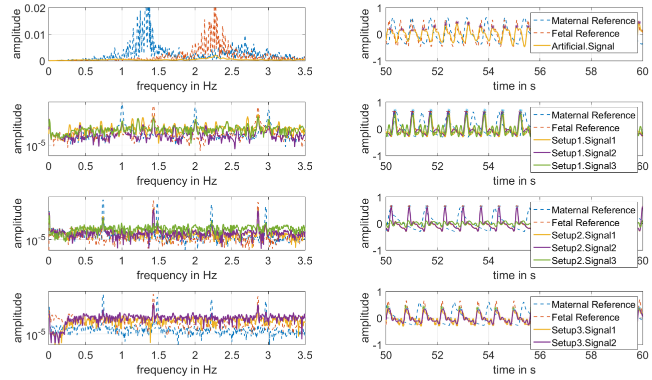

Fig. 5. The figure shows the signals after using the ANC in the frequency domain on the left and after using the comb filter in the time domain on the right. The results of the peak detector are marked in the charts on the right side.

Fig. 6 shows the resulting fetal pulse waves after averaging. In different trials 50, 20 and 10 fetal pulses were included in the averaging procedure (left to right in Fig. 6). The fetal references were taken from the pressure signal of the tube in the phantom. Although they do not exactly match the PPG, they seem to be give good orientation.

The shape of the extracted wave differs more or less from the reference depending on the measurement setup and the quality of the signal. The essential features of the pulse wave, the peaks and the minima, are detectable in all signals that are averaged from 50 pulses. If 20 or less pulses were used, noisy signals with weak fetal component show disturbed mean pulse waves. This effect is visible in Setup3.Signal2. The exact detection of the main peak is more difficult and uncertain because of its extent. A similar behavior can be seen in the Artificial.Signal. The curves are more blurred when less averages are used. The other signals have a lower

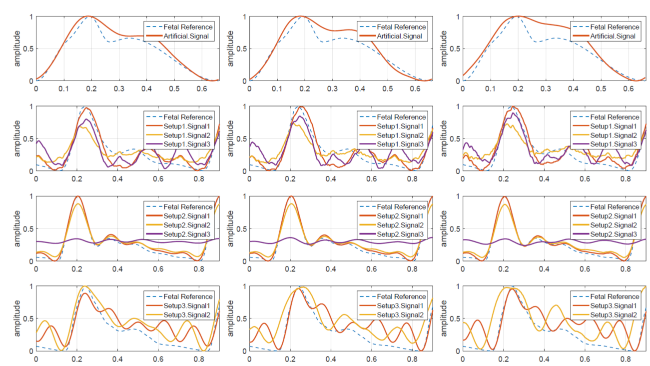

Fig. 6. The figure shows the reconstructed fetal pulse curves. From left to right, 50, 20 and 10 fetal pulses are included in the averaging procedure. The charts on top represents the artificial signal, followed by the signals measured with Setup 1, Setup 2 and Setup 3.

noise level and a relatively strong fetal component. It is obvious from Fig. 5 that the comb filtered signals already show fetal pulse waves with high quality and a good shape. This supports the peak detection and leads to an efficient averaging process. Only few averages are needed to remove the remaining noise as can be seen for example in Setup 1. The extracted pulse wave is less noisy when more averages are used, although the differences are marginal. In contrast, the number of averages does not have a visible influence to the signals in Setup 2. The shape of the pulses is nearly identical in all three plots.

IV. DISCUSSION

This work demonstrates an extended signal processing strategy for the extraction of the fetal pulse wave from mixed signals measured with transabdominal fetal pulse oximetry. The core of the strategy is an ANC and a comb filter, realized with algorithms presented in our former works [8], [7]. The application of a synchronous averaging procedure enhances the strategy to get a reliable information about the shape of the fetal pulse wave. We use synthetic signals to get a reproducible measurement environment and reference signals for the evaluation of the algorithms. A simulated signal is used to analyze the influence of a physiological pulse rate variability. Measurement noise is considered by using a tissue mimicking phantom to generate various test signals.

The results show that the ANC and comb filtered signals allow to identify single fetal pulses. In case of a low noise level, these signals might be sufficient to extract the feature points needed for pulse oximetry. However, to get reliable information about the oxygen saturation, the features need to be robust and noise invariant. We use synchronous averaging to reach this goal by suppressing the noise and the remaining maternal and mixed components. The efficacy of the approach could be shown in this work with signals that have a very weak and hardly detectable fetal component like the artificial signal and Setup3.Signal2.

The method implemented in this work also shows some drawbacks. Every inaccurate detected peak blurs the averaged curve, so that a vague detection algorithm may in worst case impair the quality of the signal. With numerous vague detected pulses, it is essential to have more pulses for the averaging procedure to compensate false positive detection an increase the SNR. For future works beyond the demonstration of the principle, it is recommended to use a finely tuned edge detector to reduce the number of false positives. In addition, it would be much more efficient to use an advanced averaging method. The quality of the resulting pulse wave would be improved if for example an a priori template matching could be used to eliminate or at least rectify the vague detections. Only reliable fetal pulse waves would be considered, so that the number of averages could be reduced by keeping equal quality.

The measurement setups used in this study tend to the ideal. We work under laboratory conditions and without motion artifacts, which is not equivalent to the reality. It is expected that it will be much more difficult to find the fetal pulse wave in measured signals from patients. Regarding real world measurements with artifacts and high noise level, the impact of a good working averaging procedure may be even more enhanced. In a next step, accuracy requirements of the features of the pulse wave should be investigated more in detail to complete the signal processing towards non-invasive fetal oxygen estimation.

V. CONCLUSIONS

A signal processing strategy for the extraction of the fetal pulse wave needed for non-invasive fetal pulse oximetry is shown in this work. We conclude that the ANC and the comb filter in combination with an averaging procedure is a feasible approach to reconstruct the shape of the fetal pulse curve. In a first step, synthetic signals were used to evaluate the algorithms and demonstrate their performance. It is shown that the averaging algorithm works well and enhances the quality of the extracted fetal pulse wave. This is crucially with disturbed and noisy PPG signals that have a weak fetal component as is expected in patient trails. To finally conclude about the suitability of the method, it is necessary to settle the requirements towards the estimation of oxygen concentrations in fetal blood. The future focus is on a more quantitative evaluation of the procedure that needs to be done in the context of pulse oximetry. Although clinical tests are necessary to finally prove our assumptions, the feasibility of the proposed averaging method is successfully demonstrated in our models.

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