Reliability of spectral analysis of fetal heart rate variability

G.J.J. Warmerdam, R. Vullings, J.W.M. Bergmans, S.G. Oei

Abstract— **Spectral analysis of fetal heart rate variability could provide information on fetal wellbeing. Unfortunately, fetal heart rate recordings are often contaminated by artifacts. Correction of these artifacts affects the outcome of spectral analysis, but it is currently unclear what level of artifact correction facilitates reliable spectral analysis. In this study, a method is presented that estimates the error in spectral powers due to artifact correction, based on the properties of the Continuous Wavelet Transformation. The results show that it is possible to estimate the error in spectral powers. The information about this error makes it possible for clinicians to assess the reliability of spectral analysis of fetal heart rate recordings that are contaminated by artifacts.**

I. Introduction

In obstetric units, it is a challenge to obtain information about fetal health. At present, cardiotocography (CTG) is the standard technique for fetal monitoring. However, the poor specificity of CTG has resulted in an increase in unnecessary operative deliveries [1]. When the fetus is at risk, too little is known about fetal health, clearly additional information is required.

Such additional information about fetal wellbeing could be provided by spectral analysis of fetal heart rate (FHR) variability. Since variations in the heart rate are regulated by the autonomic nervous system (ANS), frequency bands can be chosen such as to reflect sympathetic and parasympathetic activity of the fetal ANS [2]. Recent studies have shown that spectral powers of FHR variability are related to the fetal condition [3].

In order to ensure reliable spectral analysis, it is necessary to continuously record the FHR on a beat-to-beat basis [4]. A technique that allows for continuous monitoring of the FHR is non-invasive fetal electrocardiography (ECG), which is performed with electrodes attached to the maternal abdomen. Unfortunately, the non-invasive fetal ECG is often severely contaminated by electrical interferences such as the maternal ECG or muscle activity. As a result of the low amplitude of the fetal ECG with respect to these interferences, heartbeats can be mis-detected and the extracted FHR is frequently contaminated by artifacts.

Most of these artifacts can be detected by applying heuristic rules, that e.g. stipulate that the heart rate cannot be higher than 240 beats per minute (BPM), lower than 60 BPM, and does not fluctuate more than 20% between consecutive heartbeats. Based on the location of the artifacts in the FHR, they can be corrected for by linear interpolation between neighboring heartbeats [4]. This artifact correction will, however, affect the calculated spectral powers.

Although some studies have investigated the effect of artifact correction on spectral analysis, these studies only considered the influence of the percentage of artifact correction on spectral powers [4], [5]. Besides the percentage, the effect of artifact correction also depends on the FHR, whether consecutive heartbeats are interpolated, and the frequency band of interest. In practice, these studies thus provide little information on the error due to artifact correction. Information about this error will help clinicians to assess the reliability of obtained spectral powers. If, for example, changes in the power spectrum indicate fetal distress, it is of vital importance to know that these changes reflect actual changes in fetal ANS activity and are not the result of artifact correction.

Recently, a new technique has been developed to evaluate the spectral content of the FHR variability based on Continuous Wavelet Transformation (CWT) [6]. Since CWT uses wavelets as analytical functions, it allows for multi-resolution time-frequency analysis. Our study presents a method that uses temporal information on the location of interpolated heartbeats to estimate the error in spectral powers at each moment in time. From this, the total error in the spectral powers of a FHR recording can be calculated.

In this study, an analytical expression is derived for the error in the spectral analysis of FHR variability that is caused by artifact correction. In reality, the exact value of this error is unknown, because the true heart rate at a suspected artifact location is unknown. Therefore, a method is presented that estimates the error. To evaluate the estimation of the error, a set of FHR recordings with varying levels of artifact correction is created from artifact-free FHR recordings. Since the original FHR is now known, this approach makes it possible to quantitatively compare the theoretical error with the estimated error. Finally, the technique is applied to contaminated FHR recordings to evaluate its performance on real data.

II. Method

A. Spectral analysis

The CWT allows for multi-resolution analysis in both the frequency and time domain. Typically, a wavelet is defined as a *mother* wavelet ($\psi(t)$) with a family of *daughter* wavelets ($\psi_s(t)$). Each daughter wavelet is obtained by scaling and translating the mother wavelet. The frequency content of each daughter wavelet is associated with a certain frequency band. The CWT uses this property by comparing a certain

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daughter wavelet $\psi_s(t)$ with a signal $x(t)$. The CWT of a real discrete time FHR signal ($RR[n]$) is defined as

$$W_s[\tau] = \frac{1}{\sqrt{s}} \sum_{n=0}^{T} RR[n] \psi^*[\frac{n-\tau}{s}] \tag{1}$$

with $W_s[\tau]$ the CWT coefficients and ψ^* the complex conjugate of the wavelet. Note this study only uses real wavelets and the * indication is omitted for further calculations. From $W_s[\tau]$, the power at each time instant at a certain scale is calculated as

$$P_s[\tau] = \frac{1}{C_g} \left(\frac{W_s[\tau]}{s} \right)^2 \tag{2}$$

where C_g is the admissibility constant. The total power in a certain frequency band (P_{xF}) is then calculated by averaging $P_s[\tau]$ over time and integrating over the scales within that frequency band.

B. Error in spectral powers

The CWT in Eq. 1 can be rewritten as a convolution by defining $\widehat{\psi}_s[t] = \frac{1}{\sqrt{s}} \psi[\frac{-t}{s}]$

$$W_s[\tau] = \sum_{n=0}^{T} \widehat{\psi}_s[\tau - n] RR[n] \tag{3}$$

Since each daughter wavelet $\widehat{\psi}_s$ has a finite effective support width (sw_s), the summation over n samples is limited to the samples $n \in [\tau - \frac{sw_s}{2} : \tau + \frac{sw_s}{2}]$.

Using Eq. 3, the spectral power at scale s of the original heart rate without artifact correction (RR) and of the heart rate with artifact correction ($\widetilde{RR}$) become

$$P_s[\tau] = \frac{1}{C_g s^2} \left(\sum_j \widehat{\psi}_s[\tau - j] RR[j] + \sum_m \widehat{\psi}_s[\tau - m] RR[m] \right)^2 \tag{4}$$

$$\widetilde{P}_s[\tau] = \frac{1}{C_g s^2} \left(\sum_j \widehat{\psi}_s[\tau - j] RR[j] + \sum_m \widehat{\psi}_s[\tau - m] \widetilde{RR}[m] \right)^2 \tag{5}$$

with $j, m \subset n$, where samples j are the correct RR samples, and samples $m \cap j$ are the interpolated RR samples. Note that for samples j the value of $\widetilde{RR}[j]$ is equal to $RR[j]$.

Based on Eq. 4 and 5, the error ($\varepsilon_s[\tau] = P_s[\tau] - \widetilde{P}_s[\tau]$) is given as

$$\begin{aligned} \varepsilon_s[\tau] = & \frac{1}{C_g s^2} \Bigg[\left(\sum_m \widehat{\psi}_s[\tau - m] RR[m] \right)^2 - \left(\sum_m \widehat{\psi}_s[\tau - m] \widetilde{RR}[m] \right)^2 \\ & + 2 \sum_j \widehat{\psi}_s[\tau - j] RR[j] \cdot \sum_m \widehat{\psi}_s[\tau - m] (RR[m] - \widetilde{RR}[m]) \Bigg] \end{aligned} \tag{6}$$

By defining $A_s[\tau] = \sum_m \widehat{\psi}_s[\tau - m](RR[m] - \widetilde{RR}[m])$ and reordering the terms, Eq. 6 can be written as

$$\varepsilon_s[\tau] = \frac{1}{C_g s^2} \left(A_s[\tau]^2 + 2 A_s[\tau] \widetilde{W}_s[\tau] \right) \tag{7}$$

Similar as for the calculation of the total power in a frequency band, the total error in the frequency band (ε_{xF}) is given by time averaging of $\varepsilon_s[\tau]$ and integrating over the scales within that frequency band.

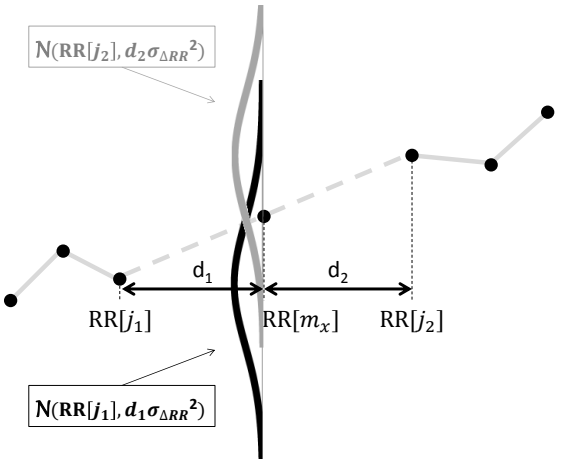

Fig. 1. Schematic illustration of the estimation of $RR[m_x] - \widetilde{RR}[m_x]$. A probability distribution is calculated for $RR[m_x]$, based on the distance between the previous correct heart rate $RR[j_1]$ and $RR[m_x]$, and on the distance between the next correct heart rate $RR[j_2]$ and $RR[m_x]$. The standard deviation of the distribution of $RR[m_x]$ is used as a measure for $RR[m_x] - \widetilde{RR}[m_x]$.

C. Estimation of the error

Since $RR[m]$ are unknown, the value of $A_s[\tau]$ is unknown in Eq. 7. Therefore, the goal is to estimate $RR[m_x] - \widetilde{RR}[m_x]$. A schematic illustration of the estimation is displayed in Fig. 1. In this image, $RR[m_x]$ is the heart rate of interest, $RR[j_1]$ is the previous correct heart rate, and $RR[j_2]$ is the next correct heart rate. The number of samples between m_x and j_1 is defined as d_1, and the number of samples between m_x and j_2 is defined as d_2.

To estimate $RR[m_x] - \widetilde{RR}[m_x]$, it is assumed that the increase or decrease in consecutive heart rates is gaussian distributed. The standard deviation of this distribution is calculated as the standard deviation of the difference of consecutive correct heart rates ($\sigma_{\Delta RR} = std(\Delta RR[j])$). The distribution in heart rates at sample m_x is then given by the product of two gaussian distributions. One distribution originating from $RR[j_1]$ ($\mathcal{N}(RR[j_1], d_1\sigma_{\Delta RR}^2)$) and the second distribution from $RR[j_2]$ ($\mathcal{N}(RR[j_2], d_2\sigma_{\Delta RR}^2)$). The standard-deviation of the distribution at x is thus given as

$$\sigma_{m_x} = \sqrt{\frac{d_1 d_2}{d_1 + d_2}} \sigma_{\Delta RR} \tag{8}$$

Because σ_{m_x} is determined by the distance from $RR[m_x]$ to its nearest correct heart rates, σ_{m_x} depends on the number of consecutive interpolated heartbeats. The value of σ_{m_x} is used as a measure for $RR[m_x] - \widetilde{RR}[m_x]$ and $A_\tau[s]$ is estimated as

$$\widetilde{A}_s[\tau] = \left| \sum_m \widehat{\psi}_s[\tau - m] \sigma_m \right| \tag{9}$$

Note there is no information available on the sign of $RR[m_x] - \widetilde{RR}[m_x]$ and $\widetilde{A}_s[\tau]$ only provide information on the amplitude of $A_s[\tau]$ rather than its sign. As a result, the estimated error is expected to be larger than the theoretical error. Only in case where $A_s[\tau]$ is predominantly positive

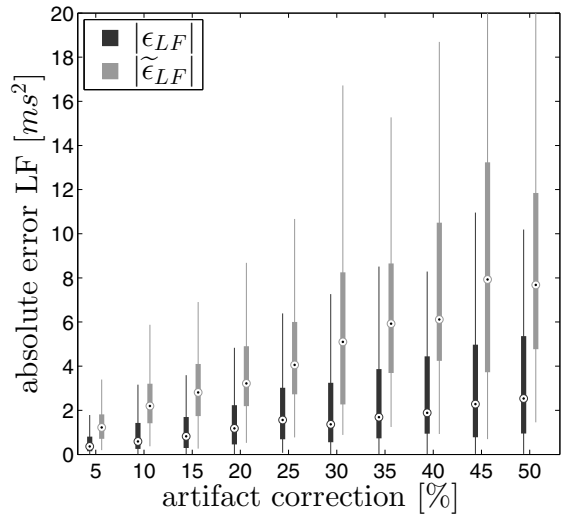

Fig. 2. The theoretical and estimated error for the LF power.

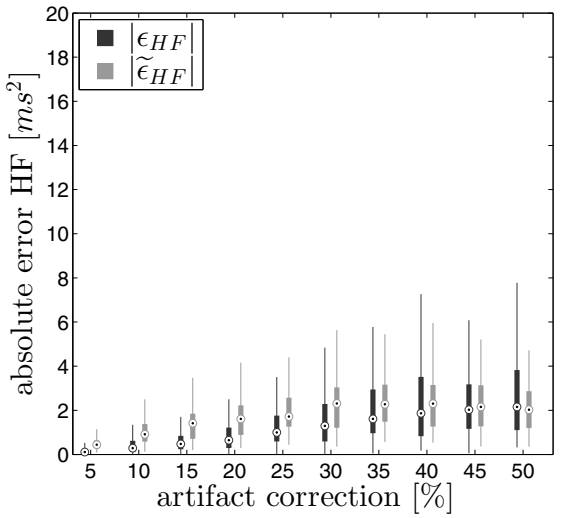

Fig. 3. The theoretical and estimated error for the HF power.

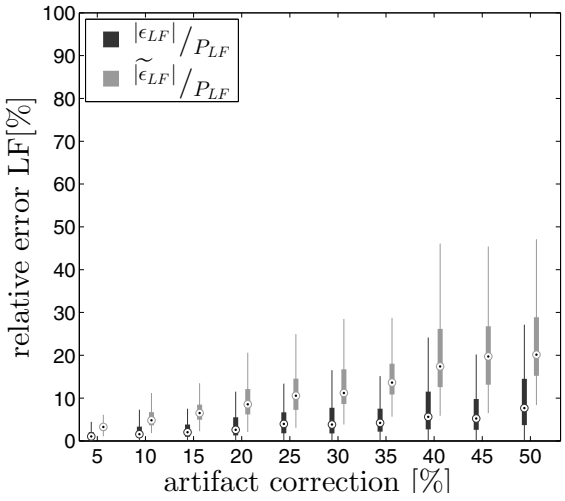

Fig. 4. The theoretical and estimated error for LF power, relative to their theoretical LF power.

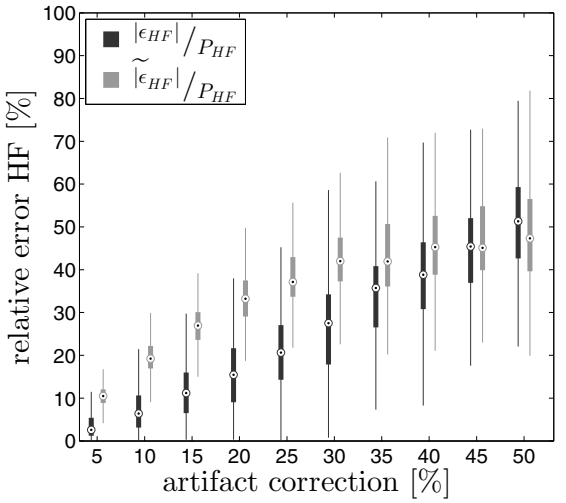

Fig. 5. The theoretical and estimated error for HF power, relative to their theoretical HF power.

and has a larger amplitude than $\widetilde{A}_s[\tau]$, it might occur that the estimated error is smaller than the theoretical error.

The error at each time instant at a certain scale is calculated as

$$\widetilde{\varepsilon}_s[\tau] = \frac{1}{C_g s^2}\left(\widetilde{A}_s[\tau]^2 + 2\widetilde{A}_s[\tau]\widetilde{W}_s[\tau]\right) \tag{10}$$

From this, the total error in a frequency band ($\widetilde{\varepsilon}_{xF}$) is calculated by averaging over time and integrating over the appropriate scales.

III. Evaluation

A. Data acquisition

A set of FHR recordings is created from abdominal fetal ECG recordings. The abdominal fetal ECG is obtained by 8 electrodes attached to the maternal abdomen. Measurements are performed at the Máxima Medical Centre (Veldhoven, the Netherlands) and are acquired at a sample frequency of 1000Hz. Visual annotation by a clinical expert is used to identify the fetal R-peaks, from which the FHR is deduced. In total, 10 artifact free FHR recordings of 10 minute length are obtained, with gestational ages ranging from 22 to 41 weeks.

Since heart rates can only be determined at times at which a heartbeat occurs, the obtained heart rates are not equidistantly distributed over time. In order to perform spectral analysis, the heart rates are first transformed into an equidistant data set by linear interpolation after which the heart rates are resampled to 4Hz [6].

Artifacts are simulated by randomly deleting heartbeats from the FHR. The artifacts are then corrected by linearly interpolated in the resampled FHR recordings. The percentages of interpolated heartbeats varies from 5 to 50%, with steps of 5%. For all artifact levels, 5 segments of 3-minutes are randomly selected from each of the original 10-minute heart rate recordings. From each 3-minute segment, 5 copies

TABLE I

RESULTS FOR TWO FHR RECORDINGS THAT ARE CONTAMINATED BY REAL ARTIFACTS.

Example	artifact [%]	LF/HF	$\widetilde{P}_{xF}$ [ms^2]	$\widetilde{\varepsilon}_{xF}$ [ms^2]
1	35	LF	70.8	125
		HF	8.9	5.1
2	6	LF	16.9	0.7
		HF	1.7	0.2

are created that each contain randomly interpolated R-R intervals, resulting in 250 unique segments to be analyzed per artifact level.

In this study the frequency bands are chosen such as to represent sympathetic and parasympathetic activity of the fetal autonomic nervous system. The low frequency band (LF, 0.04-0.15Hz), represents both parasympathetic and sympathetic activity, whereas the high frequency band (HF, 0.4-1.5Hz) only represents parasympathetic activity. For the spectral analysis of FHR variability, the fifth order symlet wavelet is selected and specific scales are used to match the frequency bands of interest.

IV. RESULTS

The LF and HF power are obtained for all FHR segments with various levels of artifact correction. The spectral powers for RR and $\widetilde{RR}$ are indicated as P_{xF} and $\widetilde{P}_{xF}$, respectively. The theoretical error is calculated as $\varepsilon_{xF} = P_{xF} - \widetilde{P}_{xF}$, whereas the estimated error $\widetilde{\varepsilon}_{xF}$ is calculated in accordance with section II-C.

Figure 2 and 3 show boxplots of the absolute value of ε_{xF} and $\widetilde{\varepsilon}_{xF}$ at each level of artifact correction for LF and HF power, respectively. Figure 4 and 5 show boxplots of ε_{xF} and $\widetilde{\varepsilon}_{xF}$ relative to their theoretical spectral power (P_{xF}). Finally, Table I shows the result of the error estimation in two FHR recordings that were contaminated by real artifacts.

V. DISCUSSION & CONCLUSIONS

Valuable information about fetal wellbeing could be provided by spectral analysis of FHR variability [3]. Spectral powers are, however, affected by artifact correction in FHR recordings [4], [5]. In order to use spectral analysis of FHR variability in clinical practice, information on the reliability of calculated spectral powers is required. This study presents a method that allows clinicians to assess the reliability of spectral analysis, based on the error in spectral powers that is caused by artifact correction.

Using the properties of the CWT, an analytical expression was derived for the error in spectral powers. The theoretical error was estimated based on information provided by the FHR. The proposed method not only accounts for differences between FHR segments, but also for interpolation of multiple consecutive heartbeats and the frequency band of interest.

In Figures 2 and 3, it can be seen that the absolute error for both LF and HF power is estimated reasonably well. Although it appears that the estimation of the absolute error is better for HF power than for LF power, this is due to the larger LF power compared with the HF power (medians 38.2 ms^2 and 4.5 ms^2, respectively). This effect becomes clear in Figure 4, where the LF error is displayed relative to the LF power. Note that, for the majority of FHR segments, $\widetilde{\varepsilon}_{xF}$ is larger than ε_{xF} and the estimation is on the safe side. The overestimation of the error is mainly because there is no sign information available on $A_s[\tau]$.

Figures 2-5 show that, besides a general increase in ε_{xF} for increasing level of artifact correction, a spread is seen in the value of ε_{xF}. The spread in ε_{xF} can be explained from differences in FHR segments. Whereas previous studies only accounted for the percentage of artifact correction, the presented method also uses information of the FHR to estimate the error. As a result, the values of the estimated error can be relatively low, even in cases where the level of artifact correction is high.

In this study, the artifact locations were randomly distributed across the FHR, which is not the case in reality. For example, if muscular activity disturbs the fetal ECG, the extracted FHR can be disturbed for multiple consecutive heartbeats. In the artifact correction simulated in this study, the effect of interpolation of multiple consecutive heartbeats is only represented to a small extent. To examine the effect on real data, Table I shows the estimated error for two real contaminated FHR recordings. The spectral information of example 1 is shown to be unreliable and the information in example 2 can be used in diagnosis. Although the effect of interpolation of multiple consecutive heartbeats should be further investigated, the results in Table I seem to be promising.

The method proposed in this study provides information on the reliability of spectral analysis of FHR. Our results show that it is possible to estimate the absolute error in LF and HF power. Further research is required to examine how this information can be used in the clinic.

REFERENCES

[1] Z. Alfirevic, D. Devane, and G M L. Gyte. Continuous cardiotocography (CTG) as a form of electronic fetal monitoring (EFM) for fetal assessment during labour. *Cochrane Database Syst Rev*, (3):CD006066, 2006.

[2] S. Akselrod, D. Gordon, J. B. Madwed, N. C. Snidman, D. C. Shannon, and R. J. Cohen. Hemodynamic regulation: investigation by spectral analysis. *Am J Physiol*, 249(4 Pt 2):H867–H875, Oct 1985.

[3] J.O.E.H. van Laar, C.H.L. Peters, S. Houterman, P.F.F. Wijn, A. Kwee, and S.G. Oei. Normalized spectral power of fetal heart rate variability is associated with fetal scalp blood ph. *Early Hum Dev*, 87(4):259–263, Apr 2011.

[4] G.D. Clifford and L. Tarassenko. Quantifying errors in spectral estimates of hrv due to beat replacement and resampling. *IEEE Trans Biomed Eng*, 52(4):630–638, Apr 2005.

[5] C.H.L. Peters, R. Vullings, J.W.M. Bergmans, S.G. Oei, and P.F.F. Wijn. The effect of artifact correction on spectral estimates of heart rate variability. *Conf Proc IEEE Eng Med Biol Soc*, 2008:2669–2672, 2008.

[6] C.H.L. Peters, R. Vullings, M.J. Rooijakkers, J.W.M. Bergmans, S.G. Oei, and P.F.F. Wijn. A continuous wavelet transform-based method for time-frequency analysis of artefact-corrected heart rate variability data. *Physiol Meas*, 32(10):1517–1527, Oct 2011.