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IMPROVING THE PERIODICITY MEASUREMENT IN FETAL HEART ACTIVITY SIGNAL

In this paper we discussed the influence of preliminary processing of the ultrasound Doppler signal on accuracy of the fetal heart rate estimation as well as on reliability of the FHR instantaneous variability assessment. We attempted to develop an optimal processing channel of US Doppler signal in order to measure the periodicity of fetal heart activity with accuracy as close as possible to that ensured by FECG. The FHR values determined from the US signal were compared to the reference data obtained from direct FECG. In a final evaluation we used the parameters describing the FHR variability as the clinically important signal features being the most sensitive to any periodicity inaccuracy. The results proved that an application of proposed algorithms improves the accuracy of interval measurements and FHR instantaneous variability assessment in relation to the new-generation fetal monitors.

1. INTRODUCTION

Electronic fetal monitoring provides a technique for assessment of uteroplacental physiology and serves to indicate the adequacy of fetal oxygenation. At present, the fetal heart activity is the only vital function of a fetus that can be recorded effectively. The most often used non-invasive method of fetal heart monitoring is the ultrasound Doppler (US) technique [6]. This monitoring relies on recording and analysis of fetal heart rate (FHR). In automated analysis of the FHR signal two main components of instantaneous variability (short- and long-term) are evaluated quantitatively by a variety of numerical indices. The short-term variability is one of the most important premises indicating appropriate fetal development, with high predictive value for automatic classification of FHR records. The early prediction of bad fetal outcomes helps to avoid dangerous situations which are impossible or more difficult to manage in the newborn. Therefore, improving the reliability of variability determination should have major influence on the quality of automated classification of FHR records. Our previous works showed that the long-term FHR variability indices derived from the US Doppler are obtained with rather a good precision [7]. However, depending on the calculation procedure, the short-term variability indices may be considerably decreased while using an autocorrelation technique to process the ultrasound Doppler wave.

The most accurate measurement of the periodicity in fetal heart activity is limited to the labour since it is ensured by acquisition of electrical signal of the fetal heart using direct fetal electrocardiography (FECG). Applying the ultrasound Doppler technique the fetal heart beats are detected from the envelope, where the Doppler shift of the reflected US wave is directly related to the speed of the heart moving parts i.e. the valves and walls.

In this paper we discussed the influence of preliminary processing the ultrasound Doppler signal on accuracy of the fetal heart rate estimation as well as on reliability of the FHR instantaneous variability assessment [3]. We attempted to develop an optimal processing channel as regards to establishing the useful bandwidth of US Doppler signal as well as the autocorrelation function parameters in order to measure the periodicity of fetal heart activity with accuracy as close as possible to that ensured by FECG. We particularly would like to answer the question if the modern US signal processing techniques allow us to measure heart cycles with the precision high enough for reliable estimation of clinically important signal features.

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2. METHODS

In the Doppler signal, the mechanical activity of fetal heart is represented as temporary increases of signal amplitude, while the signal frequency is proportional to heart valves movement speed. The useful range is comprised in the bandwidth from 50 to 500 Hz. Cutting of lower frequencies removes components usually related to fetal movements whereas the upper limit reduces blood flow effects.

The autocorrelation function (AF), when applying to analysis of the entire signal shape, requires the US Doppler envelope to be previously determined. In the basic approach the bipolar signal is converted to unipolar one by rectification, which is followed by a low-pass filtering. However, in order to obtain the most reliable periodicity measurements we determined the envelope signal by means of Hilbert transform. Determination of the autocorrelation function and the position of its dominant peak enable the measurement of the cardiac cycle duration T_i [4]. Values of T_i intervals are transformed into instantaneous fetal heart rate (FHR$_i$) expressed in beats per minute (bpm) accordingly to the equation: FHR$_i$ [bpm] = 60000 / T_i [ms].

2.1. AUTOCORRELATION WINDOW PARAMETERS

The autocorrelation function is able to determine the periodicity of US Doppler signal only if at least two corresponding mechanical activity episodes, which belong to two consecutive heartbeats, are contained in the signal window being analyzed. On the other hand, it was noticed that windows longer than two beats cause significant averaging of interval measurement and thus reduction of the short-term FHR variability [3]. Practically, as we proved in our previous work [7], only an adaptive window of the length which is set to comprise always two heart beats assures beat-to-beat accuracy of changes measurement.

If the algorithm fails to measure interval duration (which can be caused by signal interferences or low signal quality) the window length will not be updated to exactly follow the changes in instantaneous heart rate values. If the signal loss period lasts for several heartbeats, there is a considerable risk which may make the measurements impossible if the cardiac cycle duration extends rapidly, since the window becomes too short to cover two corresponding mechanical activity episodes in successive heart cycles.

As a solution we propose two-stage algorithm for accurate periodicity detection (Fig. 1). In the first stage a rough estimation of periodicity is obtained using a standard AF calculated in window of three-second length. Additionally to simplify the calculation, the envelope is downsampled to 200Hz, to reduce the required computational efforts. The period is estimated only once per second, since the only purpose is to provide information for the second – precise stage of periodicity measurement. The estimated period is used for AF window adjustment to keep its length at value of two heartbeats during the measurement process.

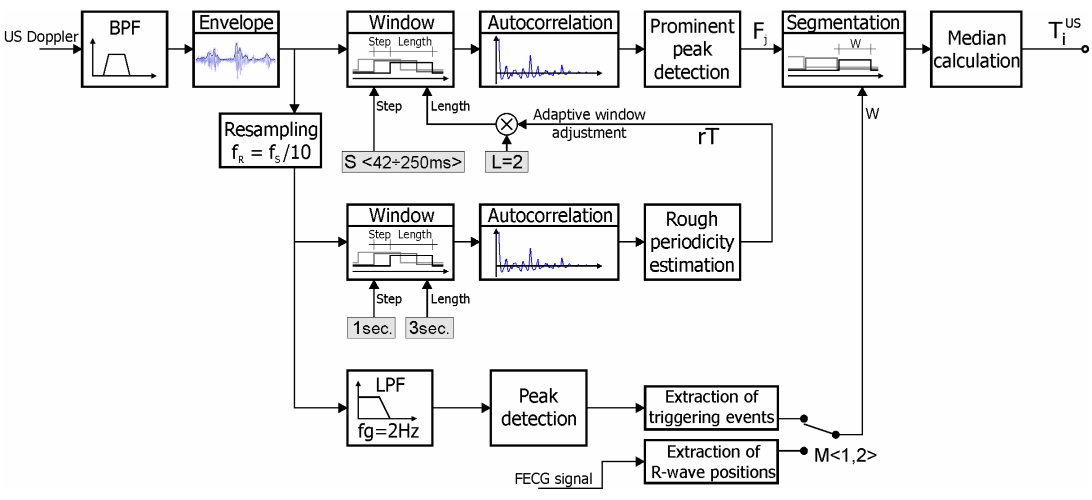

Fig. 1. General scheme of the algorithm for Doppler signal processing. The final T_i^{US} signal is a function of parameters defining the shift increment (S) and the applied method (M) for sample series F_j segmentation.

The second parameter which controls the autocorrelation function is the window shift increment, defining the time shift of a signal window between two measurements. Since the FHR variability analysis requires the each heartbeat to be represented by a single value, the maximal shift cannot exceed the duration of the shortest possible heartbeat i.e. 250 ms. However, an accuracy can be improved if the periodicity is measured several times during a given cardiac cycle. In our experiments the shift increment (S) was changing from 42 to 250 ms, which means that the number of measurements per second was between 4 and 24.

2.2. THE GAUSSIAN WINDOW

Artefacts in the US signal, usually coming from maternal blood vessels are represented in AF function as false maxima occurrences. To prevent incorrect measurements we proposed a heart cycle length prediction based on previously estimated periodicity. Then, the correction is applied to the AF using the Gaussian window, with the centre in $\tau = T_{i-1}$ (Fig. 2). Finally, that τ value is chosen as the instantaneous periodicity F_j, for which the $R_W(\tau)$ function reaches its maximal value:

$$F_j = \tau_n\{\tau_n : R_W(\tau_n) = \max(R_W(\tau))\},\ \ R_W(\tau) = R(\tau)\cdot\exp\left[-\frac{1}{2}\cdot\left(\frac{\tau - T_{i-1}}{\sigma\cdot T_{i-1}}\right)^2\right], \tag{1}$$

where:

σ stands for the parameter controlling the window width. For all further calculations the σ-factor was equal ¼.

2.3. INTERVAL VALIDATION

In order to eliminate measurement errors caused by signal interferences (i.e. related to erroneous fetal heart rate doubling or the temporary transition to maternal heart rate), a validation stage has to be applied. Because the usually used criteria [2] accept too wide range of FHR changes, we applied much more complex ones. The T_i values from the time event series should fulfill the condition:

$$T_{i-1} - 0.10\cdot\Delta_{i-1} < T_i < T_{i-1} + 0.15\cdot\Delta_{i-1}, \tag{2}$$

where:

$$\Delta_{i-1} = \begin{cases} T_{i-1} - 300ms & \text{for} \quad T_{i-1} \geq 320ms \\ 20ms & \text{for} \quad T_{i-1} < 320ms \end{cases}$$

The T_i is accepted if it belongs to the group of three consecutive intervals fulfilling the (2). The validation is carried out bidirectionally, which means that in the next step intervals are accepted according to their successors. A given interval is considered as incorrect only if it does not meet the criteria in both directions.

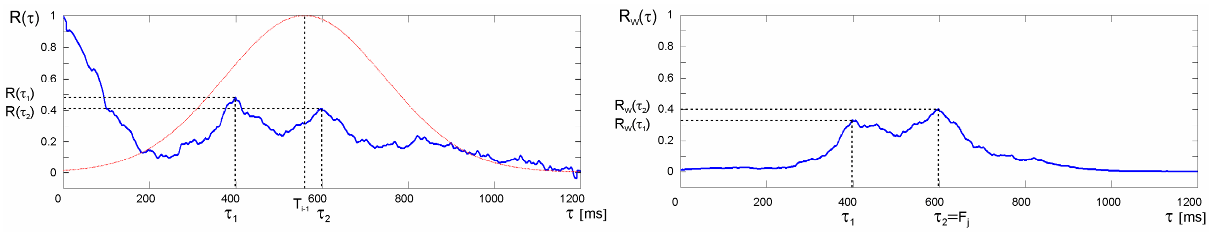

Fig. 2. The correction of the AF value using the Gaussian window. The upper plot presents AF and the Gaussian window centered according to the last correctly measured heart interval. The plot below presents the $R_W(\tau)$ function obtained after the correction of the autocorrelation function $R(\tau)$ using the Gaussian window. The correction allows to indicate an appropriate interval value τ_2 when interferences caused a false maximum - $R(\tau_1)$.

2.4. SEGMENTATION PROCEDURE

The calculation of the autocorrelation function, which is repeated many times for a given cardiac cycle, provides a new periodicity value F_j with every AF shift. Therefore, the instantaneous FHR is represented by a set of evenly spaced values. This signal has to be divided into segments corresponding to particular heartbeats and the final FHR values should be calculated from these measurement series [2]. The proposed segmentation algorithm relies on extraction of heartbeat markers from the envelope of US Doppler signal. For that reason an additional stage of signal processing is needed. It is based on the low-pass filtering of the envelope, and since the applied cut-off frequency of 2Hz [5] is very low, it requires a high order *FIR* filter. However, because for this triggering a rough estimation is enough, the envelope signal can be downsampled to simplify the computation. The pseudo-sinusoidal signal on the 60th order filter output assigns exactly one local maximum to each heart cycle. Detecting these local maxima provides rough markers of successive heartbeats – so called triggering events. For each of those events a segment corresponding to heartbeat is assigned. Finally, the value of each interval is determined as a median value of all F_j measurements contained within this segment.

The segmentation method gives the FHR signal in form of time event series, which means that each interval is described by its precisely measured duration T_i, and additionally by the marker of its occurrence in time. The position of a given interval in time is a rough estimation since it is only necessary for synchronization of T_i^{US} intervals with the reference T_i^{RR} intervals (essential to evaluate the interval measurement error of the ultrasound-based acquisition method). Having information on exact location of the heartbeats in the reference direct fetal electrocardiogram, allows us to test an influence of the AF parameters independently from our segmentation method being the last stage of the FHR determination process.

3. MATERIAL

The research material is based on two signals simultaneously recorded during a labour: fetal electrocardiogram captured directly from the fetal head as the reference signal as well as the signal from mechanical activity of fetal heart acquired using the US method. Both signals were sampled with 2 kHz. Three records of a total length of 68 minutes (8945 heartbeats) were chosen for analysis. The crosscorrelation function was used to find QRS complexes in FECG signal by comparing the signal with a template. Since the R-wave is the dominating component of the QRS complex, the crosscorrelation peak corresponds to matching of the R-waves in two complexes being compared. Therefore, a distance between two consecutive peaks is the reference T_i^{RR} interval. For quantitatively describing the short-term and long-term FHR variability we have chosen *STI* and *LTI* indices proposed by de Haan [1].

Table 1. Descriptive statistics of the reference FHR signals derived from direct fetal electrocardiography.

Record	Duration [min:s]	Mean FHR [bpm]	*STI* Mean ± *SD*	*LTI* Mean ± *SD*	Signal loss [%]	Number of heartbeats
1	45:26	132.21	9.17 ± 2.40	52.61 ± 2 6.25	3.74	6015
2	13:22	127.73	7.37 ± 0.75	18.30 ± 8.48	0.78	1188
3	9:20	130.43	6.54 ± 0.65	18.0 ± 9.44	2.09	1742

4. RESULTS

The error describing the heart interval measurement (ΔT_i) was defined as the difference between the T_k^{US} interval and the corresponding reference T_i^{RR} interval (according to the middle point of the reference interval duration):

$$\Delta T_i = T_k^{US} - T_i^{RR}, \text{ for } k \text{ meeting condition }, \tau_k^{US} \le 0.5 \cdot (\tau_i^{RR} + \tau_{i+1}^{RR}) < \tau_{k+1}^{US}. \tag{3}$$

The FHR variability indices (*STI*, *LTI*) are calculated within separated one-minute signal windows. Since their values are considerably changing between subsequent windows, the analysis of the relative variability calculation error is more suitable:

$$\delta\, STI_j = \left(STI_j^{US} - STI_j^{ECG} \right) / STI_j^{ECG}, \tag{4}$$

where *j* indicates the *j*-th minute of record being analyzed.

4.1. NUMBER OF MEASURMENTS PER SECONDS

The initial conditions of the algorithms have been set as follows: US Doppler signal bandwidth at 100÷475 Hz [5], cut-off frequency of the low-pass filter to the envelope determination at 50 Hz, number of periodicity measurement (per second) within a range from 4 to 24. Measurements of instantaneous heart rate (F_j) were transformed into time event series through their segmentation, but using an information on the exact location of heart beats in the reference FECG. Figure 3 shows that the accuracy of an interval determination increases when number of measurements decreases. This tendency is caused by determination of a given interval value as the median value from the measurements made within this interval, which eliminates boundary measurements usually responsible for the highest errors. On the other hand, the accuracy improvement that can be achieved with an increase of the shift increment is rather insignificant, whereas the computational efforts significantly increase. For further analysis, the value of 12 measurements per second was selected as a compromise.

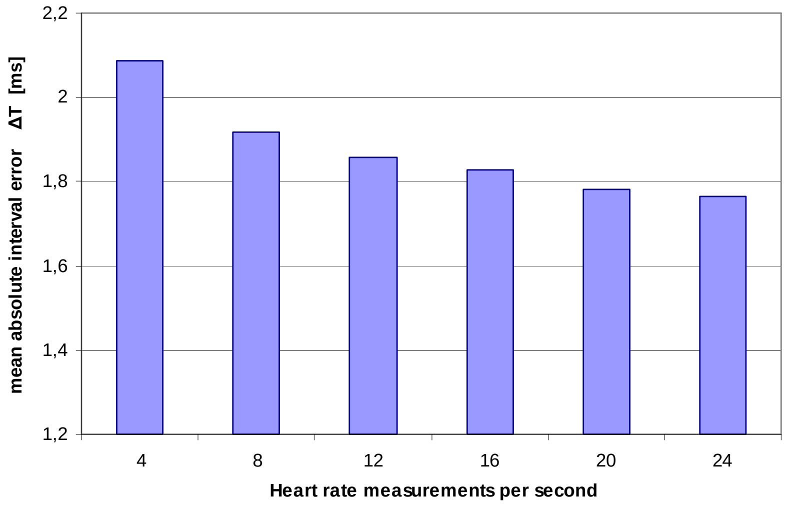

Fig. 3. Influence of the number of instantaneous heart rate measurements on the accuracy of the heartbeat duration determination.

4.2. US DOPPLER BANDWIDTH

Using the same initial conditions, the bandwidth was limited by band-pass filter of the 120-th order, whose cut-off frequencies were changing in the following ranges: 50÷180 Hz for the lower frequency and 200÷500 Hz for upper one. The obtained results are presented in Fig. 4. The best accuracy was achieved with the widest bandwidth (50÷500 Hz). However, it should be noticed that increase of the lower cut-off frequency was accompanied only by a small increase of the error value, whereas cutting of the higher frequencies caused a significant drop of the beat-to-beat determination accuracy.

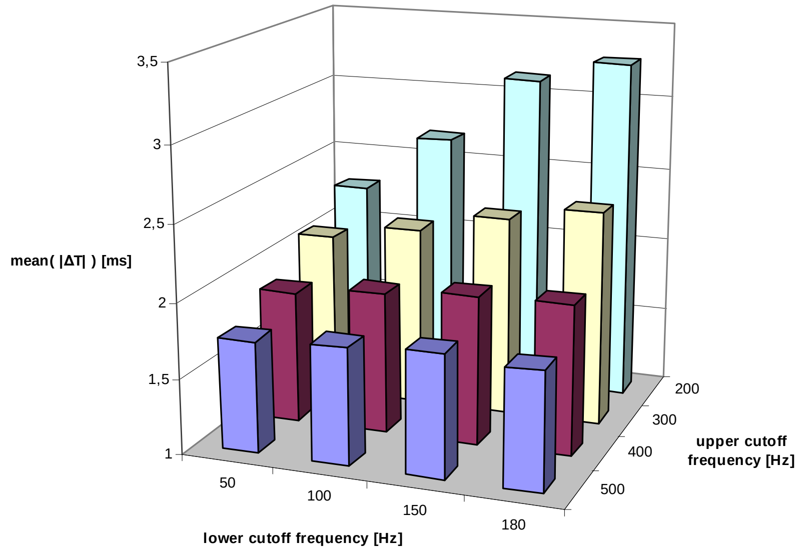

Fig. 4. The influence of the bandwidth parameters during the filtering of the ultrasound Doppler signal on the beat-to-beat determination accuracy.

4.3. ULTRASOUND DOPPLER ENVELOPE DETERMINATION

The method used to obtain the envelope of the US Doppler signal is expected to affect the accuracy of the FHR determination. The simplest method is based on conversion of the bipolar signal into the unipolar one by rectification, which is followed by low-pass filtering. The cut-off frequency is responsible for a level of signal smoothing. In our experiment the envelope was calculated four times: without any smoothing by low-pass filtering, and for the cut-off frequencies 25, 50, 100 Hz. In the next step, this simple approach was face up to the more complex one using the Hilbert transform filter, according to the formula:

$$A(n) = \sqrt{s^2(n) + \hat{s}^2(n)}\,, \tag{5}$$

where $\hat{s}(n)$ is a Hilbert transform of *s(n)* signal.

Table 2. Influence of the calculation method of the ultrasound Doppler envelope on the heartbeat measurement accuracy.

Envelope calculation methods		$\overline{\lvert\Delta T_i\rvert}$ [ms]	median($\lvert\Delta T_i\rvert$) [ms]	$\overline{\Delta T_i}$ [ms]	$SD(\Delta T_i)$ [ms]	δSTI [%]	δLTI [%]
Signal rectification		1.73	1.02	-0.16	2.73	-5.14	0.22
Low-pass filtering of rectified signal	f_C = 25Hz	1.87	1.35	-0.44	2.62	-5.96	**-0.10**
	f_C = 50Hz	1.74	1.20	-0.30	2.57	-9.49	1.15
	f_C = 100Hz	1.66	1.07	-0.20	2.57	-10.66	0.38
Hilbert transform filter		**1.57**	**0.93**	**-0.15**	**2.54**	**-4.50**	0.22

The results listed in Table 2 show that Hilbert filter ensures the best accuracy as regard both the FHR accuracy and reliability of the FHR variability description by means of *STI* and *LTI* indices. The highest error of determination of the heartbeat duration was observed for the filter with the lowest cut-off frequency. It means that too strong filtration causes the envelope peaks to be blurred.

4.4. SEGMENTATION METHOD

The extracted time event series using two segmentation methods were related to the heartbeat events obtained from the direct fetal electrocardiogram. The obtained results show noticeable advantage of the FECG-based segmentation over the approach based on extraction of triggering events from the ultrasound Doppler signal. Of course, the first approach is for verification purposes only, and it cannot be used during real fetal monitoring session, since it requires an registering of the direct fetal electrocardiogram available only during a labour.

Table 3. Influence of the segmentation method for the heartbeat measurement accuracy.

Segmentation method	$\overline{\left|\Delta T_i\right|}$ [ms]	$\overline{\Delta T_i} \pm SD$ [ms]	$\delta STI \pm SD$ [%]	$\delta LTI \pm SD$ [%]
FECG-based	1.57	-0.15 ± 2.54	-4.50 ± 18.05	0.21 ± 10.34
Extraction of triggering events	1.90	-0.21 ± 3.06	-7.61 ± 20.69	1.50 ± 11.17

5. CONCLUSIONS

In the work [3] the accuracy of FHR estimation was evaluated for MT-430 (Toitu, Japan) fetal monitor equipped with the autocorrelation-based processing algorithm implemented in hardware. The main conclusion was that new-generation fetal monitors are not capable to provide the FHR signals accurate enough for short-term FHR variability assessment. For this model of fetal monitor the authors obtained a mean absolute error equal to 2.98 ms (standard deviation of 4.18 ms). The relative error while computing the *STI* (from de Haan) variability index was equal to –39.5%. However, our new algorithms applied to signals registered with the new instrumentation developed, improve significantly the accuracy of interval measurements and FHR instantaneous variability assessment. The mean absolute error decreased to 1.90 ms (*SD* = 3.06 ms), while mean relative error of short-term FHR variability measurement was –7.61%. Further improvement is possible but it requires development of more efficient method to extract true interval values from the measurement values being a result of application of the autocorrelation function.

Testing the influence of the US Doppler useful bandwidth on the beat-to-beat determination accuracy led us to conclusion that the lower cut-off frequency can be set to ensure efficient suppression of the interferences coming from fetal and maternal movements. It does not caused significant drop of the determination accuracy. The obtained results allow us to reconsider the ultrasound Doppler method as potentially useful for reliable fetal state assessment. The results of our works on acquisition and processing of ultrasound Doppler signal are adapted in a project of mobile fetal monitor being under development. In this context the results are promising and allow us to presume that the accuracy of new-developed fetal monitor can be better than accuracy of currently available ones.

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