The influence of Doppler ultrasound signal processing techniques on fetal heart rate variability measurements

Janusz Wrobel, Janusz Jezewski, Dawid Roj, Tomasz Przybyla, Robert Czabanski, Adam Matonia

***Abstract*—** Estimation of the instantaneous variability of the fetal heart rate (FHR) is affected by the autocorrelation techniques commonly used in the Doppler ultrasound channel of today's fetal monitors. Considerably decrease of short-term variability have been noted, which is quite surprising since as it has been shown earlier, the fetal monitors determine FHR signal with quite satisfying accuracy in relation to the reference direct fetal electrocardiography. The aim of this work was to recognize a source of errors caused by the commonly used approach. The results made possible to develop the method for correction of the indices quantitatively describing the FHR variability for a given type of fetal monitor. The proposed correction relies upon cancellation of the constant error component, which has been assigned to an averaging nature of the autocorrelation function. Although the random error component remaining after correction is still not too satisfactory considering the instantaneous values, a significant improvement of reliability of the fetal heart rate variability measurement was confirmed in case of a global one-hour trace assessment.

***Keywords*—** biomedical signal processing, fetal heart rate variability, fetal monitoring, periodicity measurement.

I. Introduction

FETAL monitoring during pregnancy and labour is based on cardiotocography (CTG), which relies on analysis of three biophysical signals: fetal heart rate (FHR), uterine contraction activity (UC) and fetal movements (M). The signals are recorded by bedside fetal monitor and provided in a form of printed CTG trace which are visually accessed by clinician staff. The fetal heart rate monitoring is aimed at identification of the earliest stages of fetal hypoxia to make possible the appropriate intervention to prevent fetal asphyxia, which can result from sustained and severe hypoxia.

The FHR signal can be obtained from the mechanical or electrical activity of the fetal heart. The mechanical activity, like the opening and closing movements of valves during each cardiac cycle, may be used as events triggering a monitor when the movements can be adequately recorded and interpreted [1]. This is done by applying the Doppler ultrasound technique with beam focused on those valves. The electrical activity connected with each contraction of the fetal heart can be recorded from the maternal abdomen since the 15th week of gestation [2]. This electrical activity is significantly attenuated because the recording electrodes are not placed on the fetal heart, not even on fetal body surface – they are applied on the maternal abdomen [3]. Fetal electrocardiogram (FECG) enables recognition of heart beats, precise detection of QRS complexes and then the determination of the instantaneous fetal heart rate expressed in beats per minute (bpm), which is a reciprocal function of time interval calculated as the distance in time between a two successive cardiac cycles [4].

At present, fetal monitoring session is performed with a help of computerized fetal monitoring system. Its tasks are: analysis of data incoming from bedside monitors, dynamic presentation of signals along with analysis results as well as storing and printing the data. The system ensures access to archived records and convenient following up their longitudinal changes. The CTG signals undergo analysis aimed at extraction and quantitative description of the features essential for classification of the traces as corresponding to normal or abnormal fetal state (Fig.1)[5]. Automated analysis of the cardiotocographic signals is able to extract all the features that are hidden for visual evaluation done by clinicians (Fig.2). It is very important especially for the determination of the beat-to-beat variability of fetal heart rhythm, which is crucial for the fetal wellbeing assessment. Additionally, stable computer algorithms and threshold values significantly increase repeatability and objectivity of signals analysis. The FHR values are ranging from 60 bpm to 240 bmp, and dominating physiological values represent a range from 120 bpm to 160 bmp. The mean value of fetal heart rate calculated for one-hour record decreases with the gestational age. When classifying the fetal heart rate signals the baseline, variability and the presence of acceleration /deceleration patterns have to be assessed. The baseline FHR (the resting level of the fetal heart rate) is the fundamental pattern. Accelerations of FHR as temporary increases of fetal heart rate in response to fetal movements are a sign of fetal central nervous system alertness and fetal well-being. The temporary decreases of FHR called decelerations usually reflect such risky events as compression of the umbilical cord. During automated analysis of the FHR signal, two main components of instantaneous variability: short- and long-term

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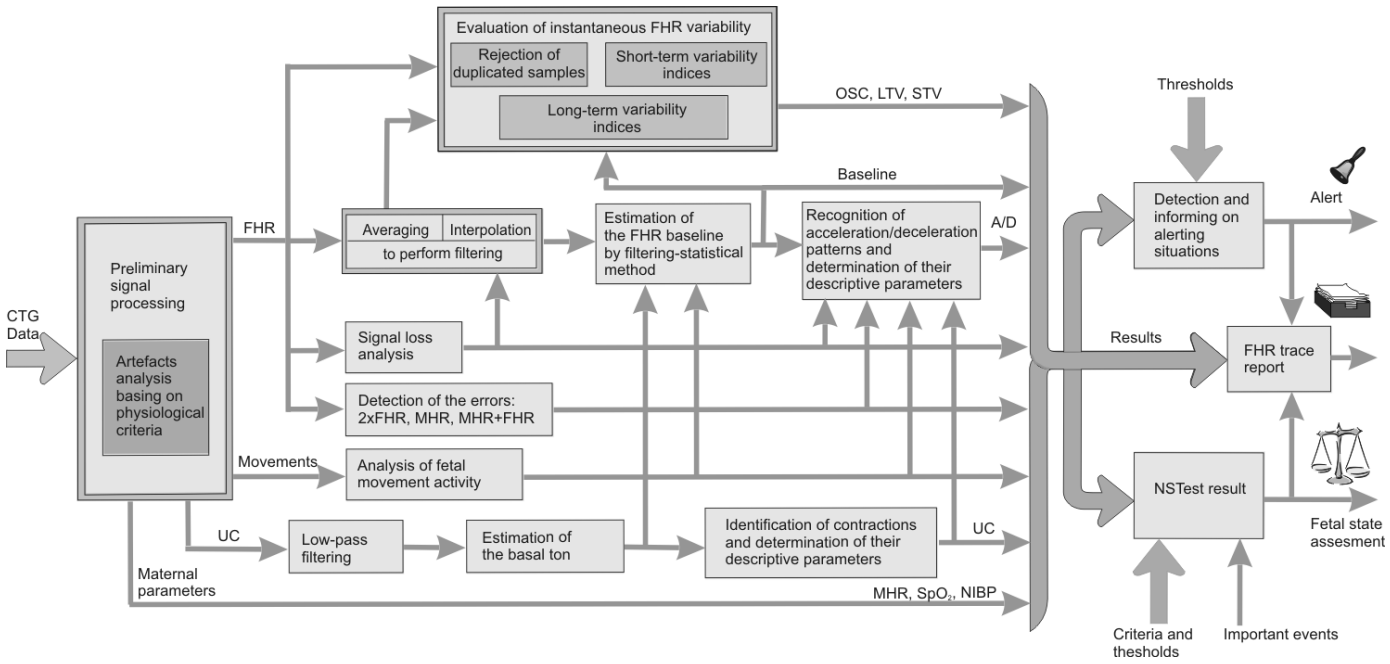

Fig.1 Block diagram of computerized analysis of fetal monitoring signals: fetal heart rate (FHR), uterine contractions (UC) and fetal movement (M).

[6], [7] are evaluated quantitatively using various numerical indices (Fig.2). They have been defined on a basis of time intervals T_i calculated between consecutive heart beats in fetal electrocardiogram (FECG) [8], [9]. However, most of present-day computerised fetal monitoring systems analyse the FHR signal which is transmitted from bedside monitors equipped with Doppler ultrasound-based technology (US) [10]. This approach affects directly the accuracy of calculation of the instantaneous FHR and thus the values of variability indices.

Dawes et al., [11] compared the ultrasound-based technique with the direct electrocardiography and noted that the error of calculation of the short-term variability (STV) index (calculated as the beat-to-beat changes) caused by the ultrasound approach reached 100 %. Lawson et al. [12] doing the same, but for other type of fetal monitor, noted a much larger calculation error (200 %). In both cases the signals were recorded using fetal monitors of former generation, which applied peak detection method to recognize fetal heart beats. In further work, using new generation fetal monitor – with an autocorrelation function – Lawson et al. [13] obtained much lower error of the STV assessment: –35 %. A minus sign means that the US approach has decreased the short-term variability in relation to the FECG.

Earlier [14], [15] we proved that the US method is able to

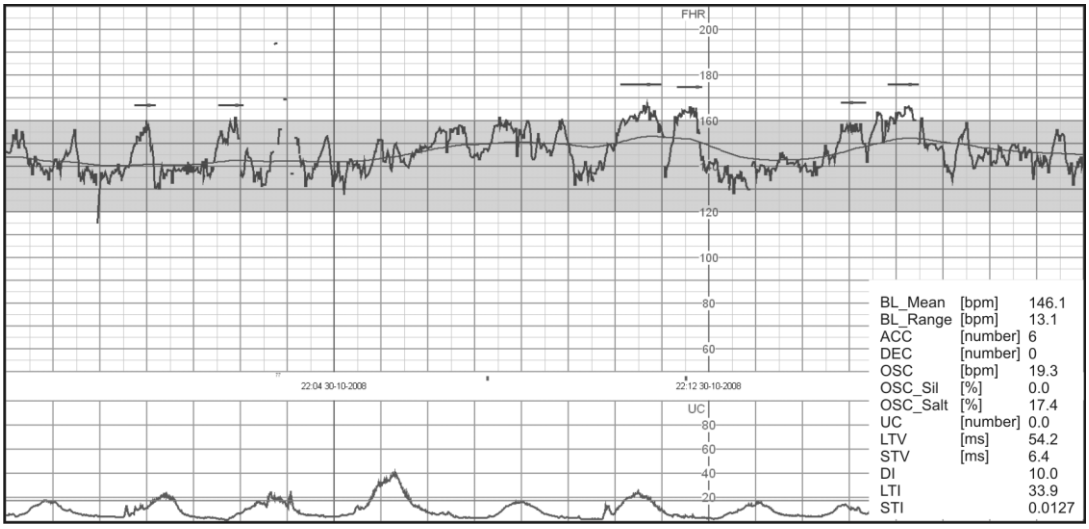

Fig.2 A screen with fetal monitoring signals: fetal heart rate with fitted baseline and marked acceleration episodes (top plot) and uterine contractions - bottom plot. Additionally, window presents 13 CTG parameters: ACC, DEC and UC represent number of patterns recognized in segment, and the OSC values – percentage of time in the segment duration.

determine T_i interval with the accuracy of about 2.5 ms in relation to the reference FECG (about 0.7 % of the typical measured T_i value of 440 ms). It was enough for both visual analysis and automated recognition of slow-wave FHR patterns: baseline, acceleration and deceleration. The long-term variability (LTV) indices were characterised by the mean error of about –5 %, which does not distort considerably the clinical assessment. However the mean error of –22 % obtained for the set of STV indices was unacceptable.

Considerable decrease of the STV indices caused by the ultrasound approach makes doubtful their use for automated clinical assessment of FHR trace. Since the computerized monitoring systems have to cooperate with the input devices being already in use we had to consider a possibility to improve the reliability of the FHR variability analysis which resulted in the new method proposed for correction of the FHR variability indices.

However there are still some questions arising. What is the reason for such significant variability decrease? Since as it has been shown, modern fetal monitors provide the FHR signal of satisfying accuracy in relation to the reference FECG considering single T_i intervals. What is the source of the US approach error, which leads to an increase of STV indices for the former generation of fetal monitors and to a decrease for the new generation? Finally, is it possible to improve the reliability of the FHR variability indices computation using the US technology, without any direct modifications of the measurement channel?

The aim of this work has been intended to answer all the above questions. In the first stage, using the collected research signal database and analyzing the results obtained earlier [14], a theoretical study on the sources of different error components in the new generation monitors – with an autocorrelation technique – was carried out. Next stages comprised several experiments to verify our expectations, and to enable the evaluation of the influence of particular error components on the beat-to-beat variability indices. Those experiments showed that the variability index error depends mainly on the heart beat intervals averaging process, being a result of autocorrelation procedures for data processing embedded in bedside monitor. At the same time a role of the measurement conditions has been found as insignificant. This enabled us to propose for a given type of fetal monitor the method for correction of the beat-to-beat variability measures.

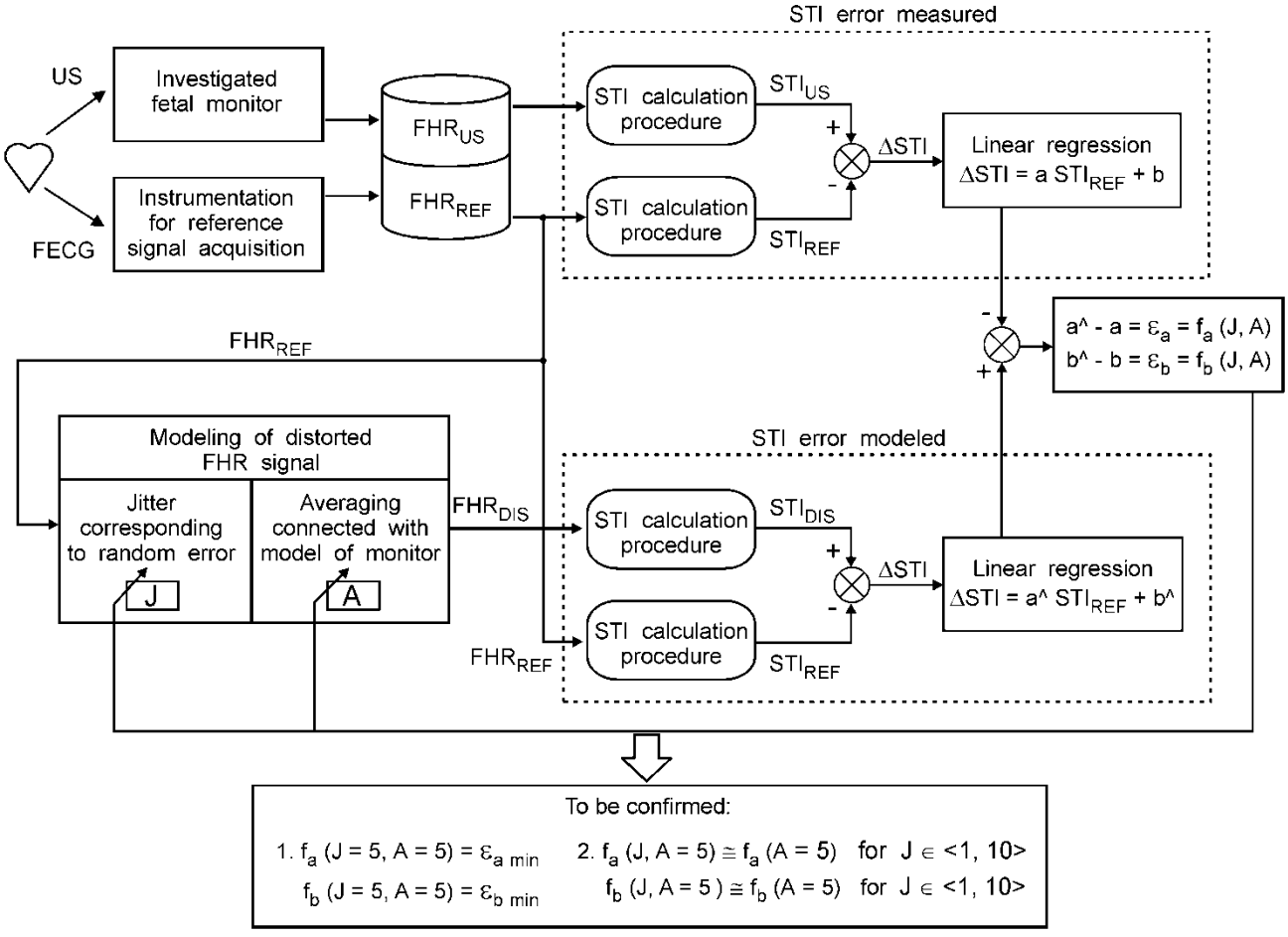

Fig.3 Overview of the methodology which comprises generally two parts. The first one is the determination of the correction parameters a and b, for the STI index measured using FHR signals collected from the given fetal monitor. The second part is the verification of the assumed error sources based on a modeling of the distorted FHR signal, which was obtained using the FHR_{REF} signal and two generated distortions: jitter (J) and averaging (A). The values of J and A were adjusted to obtain the minimum differences between the estimated correction parameters (a^, b^) and the measured ones (a, b).

Table 1 Statistical parameters of the LTI and STI variability indices determined using the US signal and the reference (REF) fetal electrocardiogram. The absolute error Δ and relative error δ of US indices are calculated in relation to the reference values.

Index	REF	US	Δ	δ [%]	Linear regression		
					Intercept	Slope	RE[a]
LTI	26,7 ± 14[b]	26,3 ± 15,2	–0,43 ± 2,6	–1,64	–0,99	0,015	2,7
STI	6,35 ± 2,6	3,88 ± 1,57	–2,47 ± 1,8	–38,9	0,93	–0,542	1,1

[a] RE – Residual Root Mean Square Error, known as standard deviation of data about the regression line,
[b] Mean ± SD

It was possible thanks to FHR signals recorded simultaneously with the use of US channel and the reference electrocardiography.

II. Methodology

In the fetal electrocardiogram recorded during labour directly from a fetal head, the QRS complexes do not change significantly from beat to beat and they can be easily recognised by detecting the evident R-waves. Whereas segments of the Doppler envelope signal, corresponding to successive heart beats are characterised by continuous change of their shape and location of the peaks [16]. Therefore, if T_i interval is determined using the peak detection method, as the distance between two consecutive heart beats, the significant location error can occur, which leads to the so called jitter error of Ti determination. This error causes an increase of one given interval and at the same time a decrease of a consecutive one. In that case the values of absolute difference between neighbouring interval durations significantly increase. It is obvious that the error is influenced by various built-in monitor features, like a sampling frequency and A/D converter resolution. This constant error component can be determined for a given monitor type, for example experimentally basing on measurement series. However, the crucial component of this error depends on some factors which are difficult to estimate because they relate to changing measurement conditions: location of the US transducer, fetal movements, stage of pregnancy development and others. Thus, in previous fetal monitors with a peak detection it was impossible to correct the computed FHR variability indices by removing the constant error component only. Considering new generation fetal monitors, the analysis of source of errors and their influence on the beat-to-beat indices has to be carried out (Fig.3) before the indices correction is proposed.

The research material comprised simultaneously recorded intrapartum fetal electrocardiograms and FHR signals from the ultrasound method. The FHR signal of 0.25 bpm resolution was acquired from RS232 output of a new generation fetal monitor. Analog FECG signal was captured directly from a fetal head and fed to data acquisition board in laptop computer. A/D sampling frequency was 2 kHz with the resolution of 12 bits. Reference FHR signal was computed

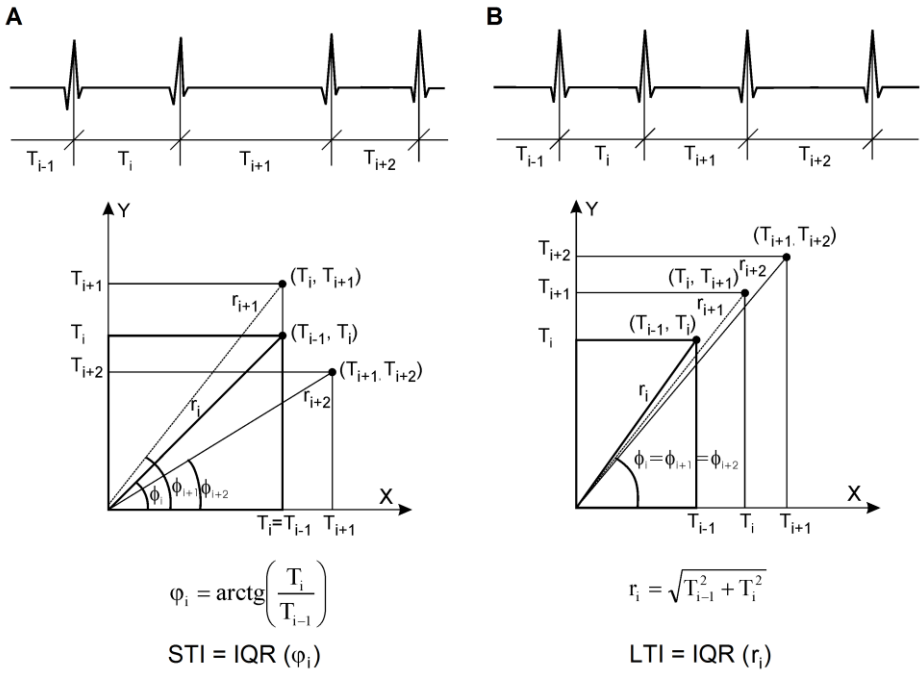

Fig.4 Graphical representation of instantaneous fetal heart rate variability: an example with dominating short-term variability – A, and with dominating long-term variability – B. The examples are illustrated using the definitions of de Haan variability indices (STI – short-term, LTI – long-term variability).

Table 2. Statistical parameters of the STI index and its error after adding to the reference signal various distortions: jitter <–1; +1> ms (J1) and <–5; +5> ms (J5), averaging over two intervals (A2), averaging over five intervals exclusively (A5), as well as adding the jitter of <–5; +5> ms and averaging (J5A5). Regression line parameters include intercept, slope and dispersion around the line (RE).

Signal distortion	STI	Δ STI	δ STI [%]	Linear regression		
				Intercept	Slope	RE[a]
J1	6,99 ± 2,5[b]	0,64 ± 0,63	9,24	0,90	–0,044	1,4
J5	12,8 ± 2,1	6,50 ± 2,03	102,0	9,55	–0,490	2,3
A2	4,61 ± 2,14	–1,70 ± 0,99	–26,8	–0,19	–0,252	0,9
A5	3,22 ± 1,54	–3,13 ± 1,57	–49,3	0,10	–0,519	1,2
J5A5	3,85 ± 1,49	–2,50 ± 1,71	–39,4	0,93	–0,547	1,3
US	3,88 ± 1,57	–2,47 ± 1,76	–38,9	0,93	–0,542	1,1

[a] RE – Residual Root Mean Square Error, known as standard deviation of data about the regression line.
[b] Mean ± SD

from the fetal electrocardiogram in an off-line mode. Two different algorithms were applied to determine T_i as the R-R intervals, and all pairs of values significantly differing were excluded. This limited the error of the reference T_i intervals to 1 ms. Finally, after rejection of the segments with signal loss, 185 minutes of traces were qualified for further analysis.

Algorithms for T_i values calculation applied in today's fetal monitors are based on autocorrelation technique [17]. These algorithms do not detect consecutive heart beats but they only estimate an averaged periodicity of the signal within a window analysed. To obtain an evident dominating peak of the autocorrelation function from noisy ultrasound signal and to consider the maximum possible beat-to-beat changes of the interval value (e.g. during the deceleration pattern), usually a window comprising three beats is used. This leads to averaging of T_i intervals, which influences the FHR variability – the STV undergoes decreasing. However, this relation is true only for indices describing linearly the FHR variability. For the most indices, which are based on nonlinear functions applied to Ti intervals, it is necessary to check how the intervals averaging affect their values.

Instantaneous variability of fetal heart rate is divided into two types. Changes concerning the durations of consecutive R-R intervals are called short-term variability or beat-to-beat variability. Due to a certain periodicity in the direction and magnitude of these changes, they result in fluctuations of the fetal heart rate around its mean level. These fluctuations are called long-term variability. Indices proposed by de Haan are the most commonly used for description of fetal heart rate variability [6]. Points corresponding to consecutive pairs of intervals (T_{i-1}, T_i) are put on two-dimensional plot in Cartesian coordinates expressed in milliseconds (Fig.4). Their polar coordinates: the radial coordinate ri and the angular coordinate φi are used to construct the definition of FHR variability indices. If short-term variability is high the angular coordinate changes from beat to beat. The domination of long-term variability is accompanied by significant changes of radial coordinate during consecutive heart beats, whereas the angular coordinate has almost the same value. The short-term variability index (STI) is defined as the interquartile range of the angular coordinates, whereas long-term variability index (LTI) as the interquartile range of the radial coordinates.

Short-term FHR variability describes the beat-to-beat changes whereas the long-term one concerns the tendency of these changes in longer time period (usually in one-minute epochs) have been defined basing on direct FECG. The

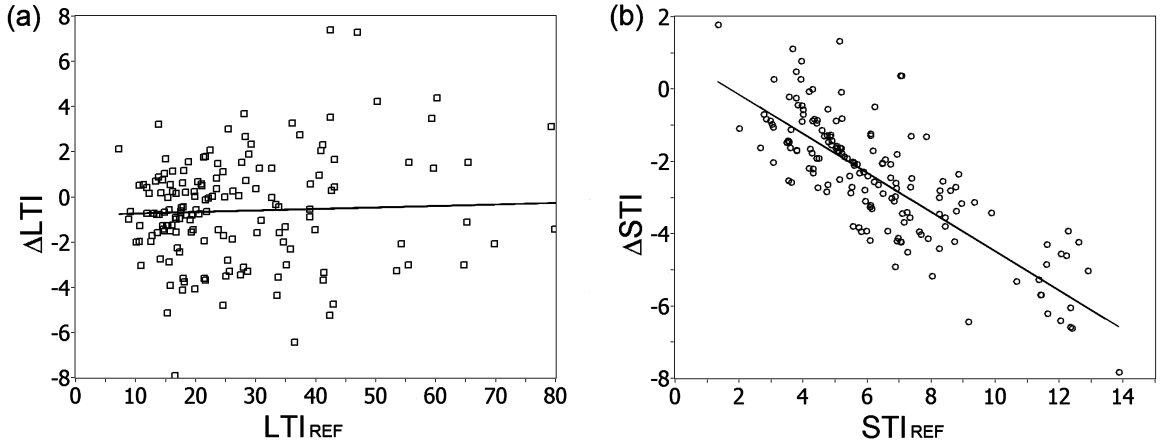

Fig.5 A scatter plot showing the relation between the absolute error of de Haan's indices: LTI for description of long-term (a) and STI for short-term (b) variability calculated as the difference between the index value from US signal and the reference (REF) one determined for fetal electrocardiogram.

Table 3. Statistical parameters of the corrected STI index and its errors, where: the indices for real ultrasound signals (US) were corrected using the regression parameters from the same signals (US*), the indices calculated for the first verification group of signals (US I) were corrected using the regression parameters from the second group (US* I/II) and vice versa – the indices from the second group (US II) were corrected using the first group (US* II/I).

Data sets	STI	Δ STI	δ STI [%]	Linear regression		
				Intercept	Slope	RE[a]
US*	6,39 ± 2,8[b]	0,00 ± 1,03	0,00	0,00	0,000	1,03
US	3,88 ± 1,57	–2,47 ± 1,76	–38,9	0,93	–0,542	1,10
US* I/II	6,09 ± 3,4	0,46 ± 1,43	8,2	0,00	0,048	1,43
US I	3,54 ± 1,47	–2,09 ± 2,77	–37,1	2,77	–0,899	1,43
US* II/I	6,11 ± 3,6	–0,11 ± 1,44	–1,7	0,00	–0,047	1,44
US II	3,31 ± 1,46	–2,91 ± 3,41	–46,8	2,78	–0,948	1,42

[a] RE – Residual Root Mean Square Error, known as standard deviation of data about the regression line.
[b] Mean ± SD

indices were calculated within one-minute segments of both the reference signal and the distorted one. Considering these definitions an averaging of Ti intervals is supposed to cause a small decrease of values of long-term indices as well as a significant decrease of short-term indices. This is illustrated in Fig.5, where obtained error values of the LTI ($\Delta LTI = LTI_{US} - LTI_{REF}$) are located much closer to zero level than the error values of STI ($\Delta STI = STI_{US} - STI_{REF}$). This error is proportional to a reference variability value which is indicated by a slope of the regression line from Fig.5b. The error originating in averaging is connected with a given monitor type with built-in autocorrelation algorithm. While the random errors from the Doppler envelope distortion they are sufficiently reduced by the autocorrelation algorithm.

Having the reference signal as the event series corresponding to the fetal heart beats detected in the FECG as well as T_i intervals determined using these events, several research experiments were performed (Fig.3). They were based on adding an artificial noise to the reference signal that allowed modeling of the real influence of the ultrasound signal processing algorithms on Ti interval values, and consequently on the STV indices. Only de Haan's STI index was selected for detailed analysis.

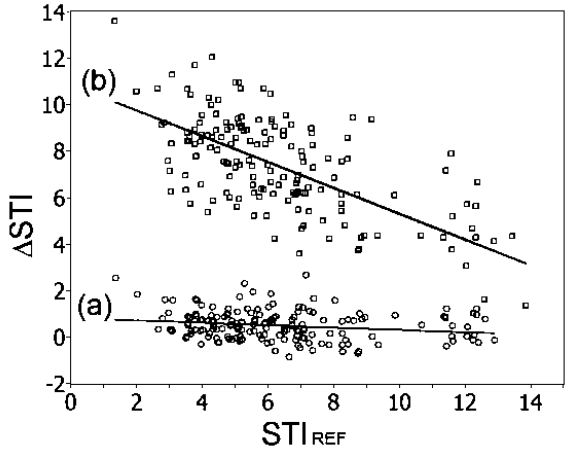

Fig.6 A scattered diagram illustrating the error of the STI index calculated for the reference signal with jitter distortion of the heart beat location of two ranges J1: <–1; +1> ms (a) and J5: <–5; +5> ms (b) against the reference index value.

In the first experiment the location distortion for the events corresponding to consecutive heart beats detected using the US method was simulated. The values from random generator of the uniform distribution were added to each reference time point defining the fetal heart beat occurrence. The random generator was run ten times with value range being changed from <–1; +1> ms to <–10; +10 > ms. In the second experiment, simulation of the error caused by averaging of the intervals in autocorrelation procedure was carried out. Reference T_i intervals from FECG were averaged over different numbers of consecutive intervals – from two to eight in each iteration. The aim of the third experiment was to simulate the entire error of the new generation monitors with various levels of jitter and averaging added. The parameters of the error dispersion (mean value, regression line) were compared with the parameters obtained for real errors of the STI index from the ultrasound approach.

The obtained results concerning a tendency of STI errors in relation to the degree of the applied distortions made possible to correct the error components appropriate for the US method, and connected with built-in autocorrelation algorithm exclusively. This correction relied on recalculation of the erroneous index values by the use of the linear regression parameters estimated for ΔSTI errors as a function of the reference values STI_{REF}:

$$\left.\begin{aligned} &\Delta STI = a \cdot STI_{REF} + b \\ &\Delta STI = STI - STI_{REF} \\ &STI^* \rightarrow STI_{REF} \end{aligned}\right\} \Rightarrow STI^* = \frac{STI}{a+1} + b \qquad (1)$$

where the STI* is the variability index corrected with linear regression parameters a (slope) and b (intercept) estimated using the reference values.

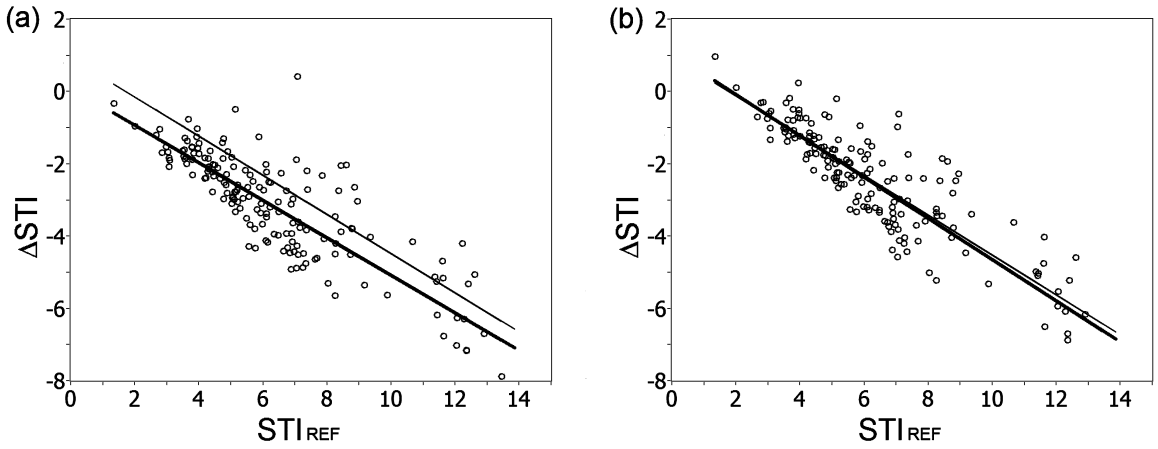

Fig.7 A plot of the error of the STI index calculated from the reference signal after its averaging (a) over five intervals exclusively (A5) and (b) after adding the jitter of <–5; +5> ms and averaging (J5A5). Additional thin lines represent the regression lines determined for the index errors for US channel.

The remaining small random error component was estimated after the correction of the STI index. Therefore, final verification of the proposed correction method had to be carried out using another database of FHR signals. They were recorded simultaneously using other type of fetal monitor based on Doppler ultrasound technology, and instrumentation for indirect fetal electrocardiography from maternal abdomen. The signals were split into two separated groups. The correction parameters estimated for one group were used to correct STI indices originally calculated for the other group and vice versa.

III. Results

It has been noted that the values of the calculation errors of STI index obtained for the research material are along the regression line (Fig.5b, Table 1) of the slope equal to –0.542 (dispersion around the line RE = 1.2). This error has a minus sign, which means that the reference value is higher than the value obtained for US, and it increases (as for the absolute value) with the index value increase. The constant component of the error is defined by the intercept of the regression line equal to 0.93.

Distorting the fetal heart beat location in time with the noise causes an increase of the value of the STV index (Fig.6). It can be seen, that this increase referring to the error component of the jitter type is disproportional to the distortion level. For the distortion with minimal range of <–1; +1> ms (i.e. 0.25 % of the most typical Ti value of about 440 ms) the index increases by about 10 %, while for the higher distortion of <–5; +5> ms (i.e. 1 % of typical Ti) its value is huge 102 % (Table 2).

In the next experiment it has been found that the averaging affects a strength of the relationship between the index error ΔSTI and the variability value STI_{REF} (Fig.7a). Negative slope of the regression line increases together with the width of the averaging window: from –0.252 for the width of two intervals to –0.519 for five intervals (Table 2). It has been appeared that the regression line slope for the averaging within five consecutive intervals is very close to the slope obtained for the ultrasound signal. Assuming the physiological mean level of FHR as 135 bpm, this range corresponds to 2.5 s. However, unlike the random error from the previous stage, the averaging itself does not lead to the positive intercept of the regression line.

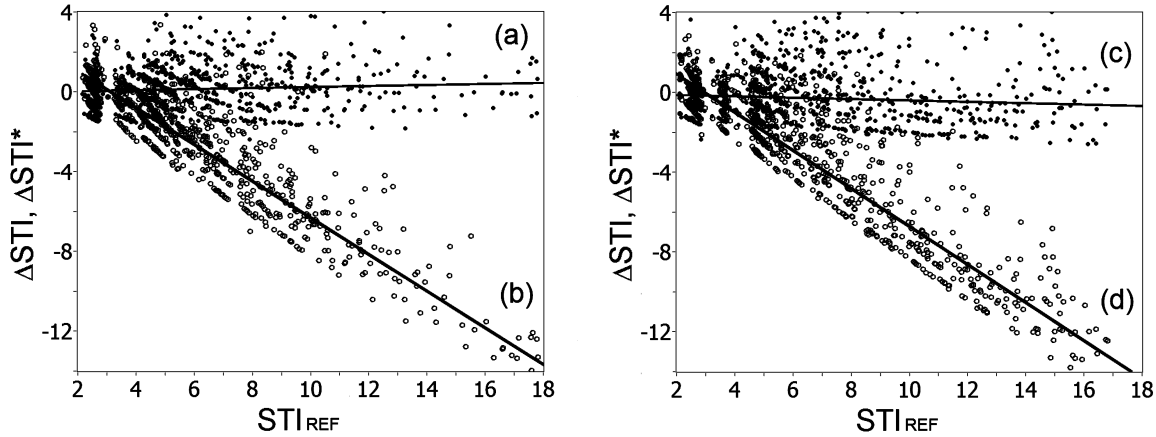

Fig.8 A scatter plot of the short-term variability index error ΔSTI* after the index correction (a). The correction has been applied using the regression line parameters determined for the errors ΔSTI (b) for the same signal database. Additionally, on the right there are histograms of the ΔSTI* (c) and ΔSTI (d).

In the third experiment, various jitter distortions as well as averaging were applied. Efforts were made to get dispersion of the STI index error as similar as possible to the results for fetal monitor (Fig.7). This was achieved for heart beat detection distortion of <–5; +5> ms and for the averaging performed within five-interval window. Statistical parameters obtained are almost identical with the real ones (compare rows J5A5 and US in Table 2). Comparing Fig.7a and Fig.7b one can see that the shift-up of the regression line is a result of adding the jitter distortion. However, in this case the distortion of <–5; +5> ms has caused the regression line intercept to change from 0.10 to 0.93 (Table 2), which means only by 0.83 without any dispersion increase around the regression line. On the other hand, the same level of jitter distortion but applied exclusively (like in the first experiment) resulted in the shift of as much as 9.55. This confirms the previous assumption that autocorrelation function decreases (by ten times in this case) the influence of the location distortion – incorrect detection of heart beats in Doppler signal – on the calculation error of the STI index. The above experiments showed that the main source of the errors of the short-term variability indices determination is the autocorrelation. Since it is strictly connected with a given type of fetal monitor, the STI index may undergo a correction for the ultrasound channel method associated with the particular autocorrelation algorithm. Such correction can be carried out using the parameters of the regression line determined experimentally using the scatter plot of relationship between the absolute determination error of a given variability index and the reference value of the index.

The last stage was aimed at verification of the efficiency of the correction performed. Values of STI index for the real US signals, were recalculated according to the regression line parameters from the same signals (US*) (Fig.8). After suppression of the constant error component, the dispersion of the differences around the zero level appeared to be uniform (RE = 1.03, Table 3) and not to have evident trend (like for LTI in Fig.5a), which indicated a pure random nature of the error. The proposed method for correction of the results of determination process of the FHR variability indices was verified basing on two additional independent groups of signals. It means that the indices calculated for the first verification group of signals (US I) were corrected using the regression parameters from the second group (US* I/II) and vice versa – the indices from the second group (US II) were corrected using the first group (US* II/I). In both cases (Table 3) the mean error for the STI index corrected was not equal to zero but it was decreased considerably: for the first arrangement from –2.09 to +0.46, and for the second one from –2.91 to –0.11 (Fig.9).

IV. Discussion and Conclusions

The detection of consecutive heart beats in time point of their true occurrence is the main problem while using the US method to acquire the fetal heart rate signal. Unlike the fetal electrocardiogram, where the dominating R-wave is easy detectable, the Doppler envelope segments often change from beat to beat, which is caused mainly by fetal movements. Imprecise detection of fetal heart beat is a main source of the error of FHR values determination in the previous generation of monitors. In the experiment carried out, where reference events – heart beats – were distorted by random noise of <–5; +5> ms value, an increase of the index values by 102 % was noted (Table 2). The application of such monitors is completely useless for analysis of the short-term variability and consequently for clinical trace assessment on its basis.

In modern fetal monitors an application of autocorrelation function for signal processing considerably limited influence of unstable shape of the US signal and its dynamic changes on the accuracy of Ti intervals determination. This technology is aimed at a precise calculation of signal periodicity instead of location of particular fetal heart beats in time. Measurement error has decreased to 2.5 ms [14], which is quite satisfactory for visual trace interpretation. However, as it has been proved the increase of Ti accuracy does not improve the analysis of the STV, because decreasing the location error influence was accompanied by adding the erroneous side effect of autocorrelation procedure – the averaging of neighbouring Ti

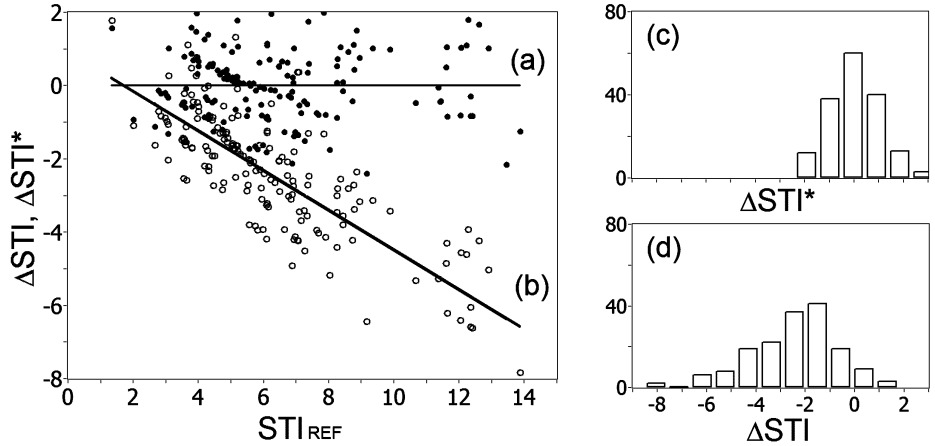

Fig.9 A scatter plot of the short-term variability index error ΔSTI* after the index correction (a). The correction has been applied using the regression line parameters determined for the errors ΔSTI (b) for the same signal database. Additionally, on the right there are histograms of the ΔSTI* (c) and ΔSTI (d).

intervals. In consequence, this leads to only a partial decrease of the mean error of STV indices calculation together with a sign change into a minus (–39 % for STI index, Table 1). In turn, the influence of the random error – heart beats location in time – goes to minimum, the regression line intercept decreased ten times (Table 2). It may be assumed, that the regression line parameters determined experimentally for a given monitor type are independent from measurement conditions, and they can be used to correct the index value for any other signal recorded by monitor of this particular type.

The proposed method for correction of the indices describing quantitatively the short-term FHR variability allow us to increase the reliability of the signal acquired from fetal monitors based on the US technology with built-in autocorrelation procedure. This correction relies upon suppression of the constant error component, which is a result of an averaging nature of the autocorrelation function. This is reflected by the ΔSTI mean value close to zero (Fig.9). The remaining random component of the averaging process reached value RE = 1.44. It corresponds to 24 % of typical value of STI which is about 6.0 (Table 3). This ensures the improvement of global evaluation of FHR variability for the entire patient's monitoring session, whose duration is usually 60 minutes, because the final variability is computed as a mean value over particular one-minute variability values.

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REFERENCES

[1] M. I. Ibrahimy, M. B. I. Reaz, M. A. Mohd Ali, T. H. Khoon, A. F. Ismail, "Development of an efficient algorithm for fetal heart rate detection: A hardware approach", *Proceedings of the 5th WSEAS Int. Conf. on Instrumentation, Measurement, Circuits and Systems*, China, pp. 12-17, April 16-18, 2006.

[2] T. Kupka, J. Jeżewski, A. Matonia, D. Roj, K. Horoba, "Adaptive processing of bioelectric abdominal signals to improve the reliability of fetal home telemonitoring", in *Latest Trends on Computers*, Editors: N. Mastorakis, V. Mladenow, *Recent Advances in Computer Engineering Series*, WSEAS Press, pp. 196-201, 2010.

[3] J. Jezewski, T. Kupka, K. Horoba, R. Czabanski, J. Wrobel, "A Problem of Maternal and Fetal QRS Complexes Overlapping in Fetal Heart Rate Estimation", in *Selected Topics in Applied Computer Science*, Editors: Hamido Fujita, Jun Sasaki, *Applied Computer Science Series*, WSEAS Press, pp. 122-127, 2010.

[4] M. I. Ibrahimy, M. B. I. Reaz, M. A. Mohd Ali, T. H. Khoon, A. F. Ismail, "Hardware Realization of an Efficient Fetal QRS Complex Detection Algorithm", *WSEAS Transactions on Circuits and Systems*, Vol. 5, Issue 4, pp. 575-581, April 2006.

[5] M. Jeżewski, R. Czabański, D. Roj, T. Kupka, "Influence of input data modification of neural networks applied to the fetal outcome classification", in *Latest Trends on Computers*, Editors: N. Mastorakis, V. Mladenow, *Recent Advances in Computer Engineering Series*, WSEAS Press, pp. 202-207, 2010.

[6] J. De Haan, J. H. van Bemmel, B. Versteeg, A. F. L. Veth, L. A. M. Stolte, J. Janssens, T. K. A. B. Eskes, "Quantitative evaluation of fetal heart rate patterns I. Processing methods", *European Journal of Obstetrics and Gynecology*, Vol. 3, pp. 95-102, 1971.

[7] E. H. Hon, "Instrumentation of fetal heart rate and fetal electrocardiography. II. A vaginal electrode", *American Journal of Obstetrics and Gynecology*, Vol. 86, pp. 772-778, 1963.

[8] H. P. Van Geijn, "Analysis of heart rate and beat-to-beat variability, interval difference index", *American Journal of Obstetrics and Gynecology*, Vol. 138, pp. 246-252, 1980.

[9] S. Y. Yeh, A. Forsythe, E. H. Hon, "Quantification of fetal heart rate beat-to-beat interval differences", *Journal of Obstetrics and Gynecology*, Vol. 41, pp. 355-363, 1973.

[10] J. Jezewski, K. Horoba, J. Wróbel, A. Matonia, T. Kupka, "Adapting new bedside instrumentation for computer-aided fetal monitoring using efficient tools for system reconfiguration", in *Recent Advances in Computers, Computing and Communications*, Editor: N. Mastorakis and V. Mladenov, WSEAS Press, Greace, pp. 80-85, 2002.

[11] G. S. Dawes, G. H. A. Visser, J. D. S. Goodman, C. W. G. Redman, "Numerical analysis of the human fetal heart rate: The quality of ultrasound records", *American Journal of Obstetrics and Gynecology*, Vol. 141, pp. 43-52, 1981.

[12] G. W. Lawson, G. S. Dawes, C. W. G. Redman, "A comparison of two fetal heart rate ultrasound detector systems", *American Journal of Obstetrics and Gynecology*, Vol. 143, pp. 840-842, 1982.

[13] G. W. Lawson, R. Belcher, G. S. Dawes, C. W. G. Redman, "A comparison of ultrasound (with autocorrelation) and direct electrocardiogram fetal heart rate detector systems", *American Journal of Obstetrics and Gynecology*, Vol. 147, pp. 721-722, 1983.

[14] J. Jezewski, J. Wrobel, K. Horoba, "Comparison of Doppler ultrasound and direct electrocardiography acquisition techniques for quantification of Fetal Heart Rate variability", *IEEE Transaction on Biomedical Engineering*, Vol. 53, pp. 855-864, 2006.

[15] J. Jezewski, T. Kupka, K. Horoba, "Extraction of fetal heart rate signal as time event series from evenly sampled data acquired using Doppler ultrasound technique", *IEEE Transaction on Biomedical Engineering*, Vol. 55, pp. 805-810, 2008.

[16] S. A. Shakespeare, J. A. Crowe, B. R. Hayes-Gill, K. Bhogal, D. K. James, "The information content of Doppler ultrasound signals from the fetal heart", *Medical and Biological Engineering and Computing*, Vol. 39, pp. 619-626, 2001.

[17] M. Y. Divon, "Autocorrelation techniques in fetal monitoring", *American Journal of Obstetrics and Gynecology*, Vol. 151, pp. 2-6, 1985.