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Fetal Breathing Transmission in Phonocardiographic Monitoring Telemedicine Systems

F. Kovács[1], Ph.D., Cs. Horváth[2], M. Török[3], Ph.D., G. Hosszú[2], Ph.D.
[1] Pázmány P. Catholic University, Budapest, Hungary
[2] Budapest University of Technology and Economics, Budapest, Hungary
[3] OGYIK State Medical Centre, Budapest, Hungary

Abstract – **The phonocardiographic monitoring of fetal heart activity due to its passive nature enables extremely long measuring times providing thus the possibility of fetal breathing recovery. The long monitoring time is required because of the temporary appearing of breathing movement. However, the long measurement time and consequently the large amount of data to be transmitted to the hospital's computer centre may be costly on mobile phone network. To keep monitoring costs low an appropriate data compression should be applied assuring the transmission of all important features of the detected acoustic signal. The present work summarizes the results of the extension of the novel telemedicine system [1] with measurement of breathing periodicity and the achieved compression level of acquisited data. The Golomb-Rice compression is applied for lossless transmission of the segmented beat cycles considering the importance of the given segments in order to obtain the most accurate transfer of beat-to-beat time and all irregular heart sounds.**

I. Introduction

Phonocardiography is a suitable tool to test fetal well-being having been studied for years by many research laboratories, mostly due to its passive nature [2][3]. Using this approach, long-term measurement of the fetal heart rate (FHR) and collection of heart sound artifacts becomes possible. The widespreadly accepted ultrasound Doppler process is not suitable for taking measurements over extended periods of time.

Our research group started from the well-known fact that certain data, eg. fetal breathing can be recorded by long-term measurement only. A preferred solution is the use of home monitoring utilizing the passive phonocardiogaphic method.

A well-known method of FHR determination is the autocorrelation, capable of determining the time of the beginning of the systolic and diastolic period (S1 and S2 sounds) even at high disturbances. Identifying these sounds the beat-to-beat time T_{bb} and its variability can be calculated. Furthermore, the additional (irregular) fetal sounds as the recoil sound, murmurs etc. can also be identified. Precise determination of T_{bb} timings is crucial, as the performance of electronic fetal monitoring methods based on computer-assisted CTG evaluation is improved when the input FHR signal is accurate and has a high resolution [4][5].

In some time periods, the acoustic signal recorded on the maternal abdomen is very disturbed influencing the shape of the S1 and S2 sounds. To overcome this difficulty the correlation method is applied on two frequency bands exploiting the different level of distortion on these [6]. The data obtained this way satisfies diagnostic demands and thus can be used for fetal monitoring. During the analysis of the records has shown that in addition to FHR determination, information on the fetal breathing might be recovered. This could mean that breathing movements are possibly recoverable without the use of additional sensors or ultrasound imaging. However, because of the temporary appearance of the breathing movement this measurement requires sometimes extremely long monitoring time.

This long-term monitoring can be performed at home by a telemedicine system. Such a system has been published in [7], however, since it utilizes the ultrasound technique, its long-term data acquisition capabilities are limited. A more suitable solution is provided by the phonocardiography applied thus in our telemedicine system [8].

This paper focuses on the specific problems concerned on the transmission of large amount on data required to detect and measure fetal breathing movement. This amount of data transmitted to the hospital's centre must be kept low, therefore a suitable data compression method is required. This compression must be lossless in order to transfer accurate beat-to-beat timings and all irregular sounds. Different compression and encoding methods have been tested and the results examined.

II. Methods

The acoustic signals are received by the narrow-bandwidth electrodynamic sensor that provides a sufficient suppression against external noises as well as maternal heart activity [8]. According to the certainty of S1 and S2 sounds identification three types of time periods can be distinguished.

Previous implementation of the system depended on a simple method to decide whether a sound cycle should be transmitted or not. When the sounds were well identifiable, only the calculated T_{bb} had to be transferred. If no sounds could be reliably identified even from the cross-correlation of four sequential heart cycles, the entire cycle was transmitted in an uncompressed form. Practical experience showed that this keeps the amount of transmitted data at an unacceptably high rate.

A suitable resolution could be the segmentation of the sound cycle and the decision whether the data should be transmitted is performed on every segment, thus expectably resulting in lower transmission rates.

A. Breathing and its influence on T_{bb}

Breathing is a very important feature of fetal well being investigated widely in the last years [9][10]. Several approaches have been taken to extract the fetal breathing rate using the variation of the FHR curve. An extensive investigation was carried out by [11], where correlation between the increase of signal intensity and breathing was experienced.

A second effect is the increase of the variability of T_{bb} during the active period compared to the quiet state, measured exactly by [12]. The increase in systolic time interval variability was also observed. That obviously requires a very accurate determination of heart sound timings. The third possibility for identification of breathing is that the most probable time length of the breathing movement of normal fetuses is 1.1-1.3 sec [13][14].

Upon the calculation of the T_{bb} values, the variance of the vector formed from these elements is determined. However, an adaptive limit is defined to filter T_{bb} values in order to diminish the effect that unsuccessfully evaluated segments, like lost heart sounds, would imply on the T_{bb} vector. As shown on equation (1), a small negative number noted as ε is assigned to elements where T_{bb} would be greater than the adaptive limit δ.

$$\Delta_i = \begin{cases} \left|T_{bb_{i+i}} - T_{bb_i}\right|, & \left|T_{bb_{i+i}} - T_{bb_i}\right| < \delta \\ \varepsilon, & \left|T_{bb_{i+i}} - T_{bb_i}\right| \geq \delta \end{cases} \qquad (1)$$

The method to calculate and filter the T_{bb} elements is presented on Figures 1 and 2.

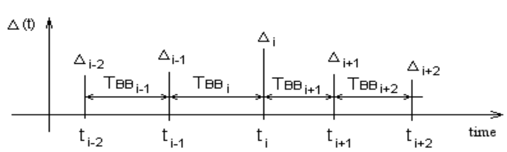

Fig.1. Calculating T_{bb} variance

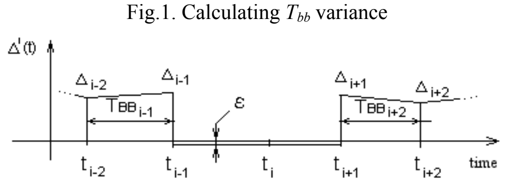

Fig.2. Interpolation and filtering in T_{bb} variance calculation

In order to indicate breathing movements, the received signal was low-pass filtered with 25Hz cutoff frequency and an intensity curve formed by integrating for an appropriately selected time window with one sample overlap. In many places of the disturbed intervals significant periodicities were found, that, according to the cited publications may be the results of fetal breathing. To separate the breathing from other fetal movements or even disturbances having similar characteristics the following rules were applied to reliably identify the breathing: 1) minimum five 1.1-1.3 sec long periodicities should be appear successively, 2) during these time intervals the variability should be significantly higher than in undisturbed periods.

B. Splitted signals and sound artifacts

Several types of heart-originated disturbances exist in the recorded signal. Our main concern here is to detect suspicious murmurs and deformations in the main heart sounds. An example of this kind of deformation is that in some cases in the S2 sound the sounds of the closure of aortic and pulmonary valves divide and the so-called split signal appears that can be described by the equation (2),

$$S(t) = A_a(t)\sin(\varphi_a(t)) + A_p(t - t_0)\sin(\varphi_p(t)) \qquad (2)$$

where $A(t)$ and $\varphi(t)$ are the time dependent amplitude and phase and t_0 is the splitting time. This phenomenon was intensively investigated in [15] using wavelet transformation to decompose the signal. However, due to timing constraints, in our method the timing of the S2 sound will always be defined by the first signal peak that corresponds normally to the closure of the aortic valve. In this case, the entire heart sound segment is transferred to the computer centre where exact analysis of the split can be performed.

C. Segmentation of the heart sound cycle

The segmentation algorithm should break up the sound cycle into four parts, namely the S1 (named *Seg1*) and S2 (named *Seg3*) sounds, murmurs and extra sounds between S1 and S2 (named *Seg2*), extra sounds and resting period between S2 and the S1 of the following beat (noted as *Seg4*). Figure 3. shows the segments with their respective numbers.

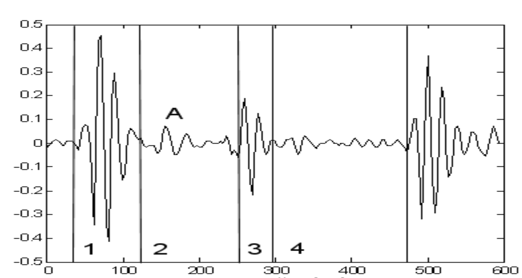

Fig.3. Segmentation of a heart sound cycle

The segmentation algorithm is based upon the calculation of autocorrelation between following cycles and applying an averaging method. The sound cycle and the result of the correlation are then transferred into frequency domain by Hilbert transformation. Previous studies describe that an approximate identification of the S1 and S2 sounds based on frequency domain comparison is possible. Based upon the transformed correlation a measure of importance is defined describing whether a particular area of the signal is considerable (eg. containing a heart sound or a suspicious murmur).

When the measure of importance is calculated, it is possible to decide what information should be transferred. The decision is carried out as follows:

1) relatively noiseless sound cycle with well identifiable sounds and no artifacts: only the cycle length and the calculated T_{bb} value needs to be transmitted;

2) detected artifact in S1 or S2 sounds (eg. split sound is present) but no murmurs are reported by the analysis: both *Seg1* and *Seg3* are transmitted entirely, but the other segments are not;

3) extra sounds or suspicious murmurs are detected in *Seg2* or *Seg4*, using intensity function analysis: *Seg1*-*Seg3* are transmitted entirely; *Seg4* is divided into two subsegments and further analysis is performed deciding whether both *Seg4* subsegments are to be transmitted or not.

Another point of interest is the presence of unusual signals in sound segments *Seg2* and *Seg4*, since these usually note extra heart sounds. Sequential *Seg2* and *Seg4* segments are compared to detect sudden changes in sound intensity therefore resulting in interception of possible artifacts. Segments are also compared to predefined intensity templates to be able to detect murmurs that are constantly present in the signal and therefore would not be identified by the sequential comparison.

D. Application of the Golomb-Rice compression

The introduction of the recovery algorithm of irregularities increases the amount of data to be transmitted, especially when the measuring unit does not have the computational capacities to perform a complex analysis of the recorded signal. Several compression methods have been evaluated, including ones that are commonly used, eg. LZW. The Golomb coding is an entropy coding that is optimal for geometric distribution alphabets. As shown on equations (3) and (4), parameter b is used to divide the input value x, forming the quotient q and the remainder γ. Quotient is unary encoded, whereas the remainder is truncated binary encoded. The parameter b is a function of the corresponding Bernoulli process; p and b are related by equation (5). In Rice coding the parameter b is a power of 2.

$$q = \left\lfloor \frac{(x-1)}{b} \right\rfloor \tag{3}$$

$$\gamma = x - qb - 1 \tag{4}$$

$$(1-p)^b + (1-p)^{b+1} \le 1 < (1-p)^{b-1} + (1-p)^b \tag{5}$$

Since the baseband heart sound signal is not a source with geometric distribution, 10th-order linear prediction has been applied to the signal, and only the error function has been encoded using the Golomb-Rice method. Adaptive Golomb-Rice compression, with source transformation has also been evaluated.

E. Description of the complete system

The telemedicine system consists of two parts, a low-cost battery-operated home processing unit providing basic signal processing capabilities and the computer centre capable of detailed analysis and processing of highly disturbed periods.

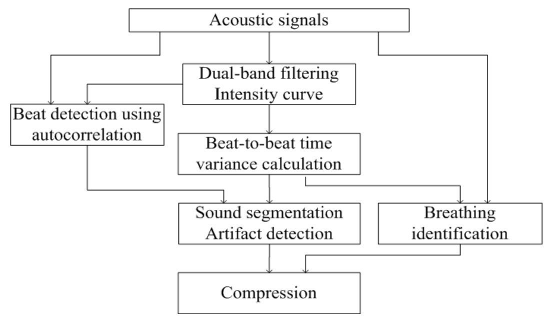

Fig.4. Flowchart of the processing system

The first step is the separation of the two frequency bands B_I and B_{II} with fifth order Butterworth digital filtering. Following this, the autocorrelation is carried out partitioned into sections according to the predicted S1 and S2 sounds. Depending on these the FHR, T_{bb}, baseline and the short- and medium-term variability can be calculated. If the confidence level of the results obtained from the prediction is lower than a predefined limit, the correlation is repeated over a 4-beats long interval in order to find the dominant heart sounds. If the repeated attempt is still unsuccessful, the full interval is transferred to the computer centre.

Having the correct timing of the S1 and S2 sounds the next step is the search for additional irregular fetal sounds. The rule applied here was that at minimum five successive appearances are required for the phenomenon to be accepted as an irregularity.

The output data of the processing is collected temporarily in the local memory and transferred to the computer centre via GPRS. There the processing is completed and permanent surveillance of even 32 patients is performed.

III. Results

A total of several thousand phonocardiographic records have been taken from more than one thousand patients in the last seven years. For extensive research of fetal breathing movements, 35 patients with gestational age of 36-40 weeks have been selected. Records with extremely low received signal or very strong maternal heart beat sounds were rejected. The global hit rate of S1 and S2 sounds for all these records was over 95%. The 3 weeks measurement of the patients happened at home. The average measuring time was 40 minutes, but at 10 patients, it exceeded 1.5 hours in order to obtain more data for fetal breathing recovery. All measurements were done in the morning between 9:00 and 12:00 in order to observe at maximum breathing activity. In selected cases, breathing movements were simultaneously checked by 2D ultrasound imaging. The majority of the problems in breathing recovery originated from maternal pulse and fetal trunk and limb movements. In spite of these, a large number of breathing episodes have been detected, most of them confirmed by the ultrasound visualization. An example of the influence of fetal movements on the T_{bb} values is shown on Figure 5. Markers (1) show epochs with unreliable signals. The T_{bb} variance is presented with different averaging window lengths (markers 2 and 3). External markers (4) show where the supervising obstetrician detected breathing movements.

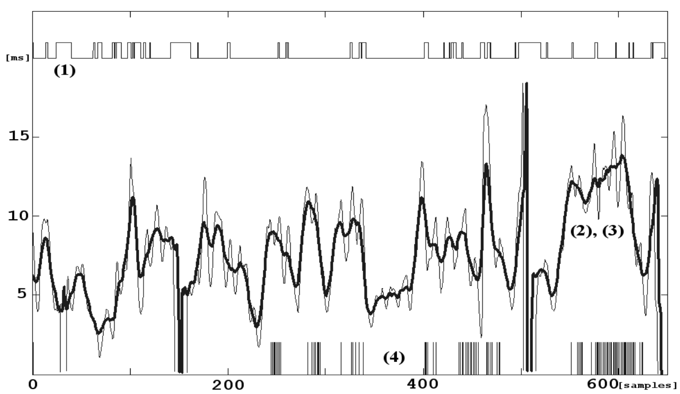

Fig.5. Effect of breathing movements on T_{bb} variance

The developed artifact detection algorithm has proven to be successful, as a case with atrial septal defect has been discovered. The corresponding murmur in *Seg2* is marked by the "A" sign in Figure 3.

Based upon a test with 300 records using two types of Golomb-Rice (GR) encoding and applying linear prediction, the ratio of the encoded data to the total record lengths are shown on Table 1.

Type of data compression	Original GR	Adaptive GR	GR with linear prediction
Average level of compression	73 %	70 %	54 %

Table 1. Encoding ratios

IV. Conclusion

The results verified that a phonocardiographic telemedicine system could be built up for FHR measurement and breathing recovery. A large number of breathing episodes have been detected, but further efforts to increase the reliability of the method are required. The measured periodicity of the breathing may be an indicator of fetal well being. This system can be used for the long-term home surveillance of pregnancies.

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References

[1] F. Kovács, Cs. Horváth, M. Török, G. Hosszú: "Long-Term Phonocardiographic Fetal Home Monitoring for Telemedicine Systems", *Proc. 27th Ann. Intern. Conf. IEEE Eng. Med. Biol. Soc., EMBC'05,* Shanghai, 1-4. Sept. 2005, p.313.

[2] N. Colley, D. G. Talberg, D. P. Southall, "Biophysical Profile in the Fetus from a Phonocardiographic Sensor", *Eur. J. Obstet. Reprod. Biol.*, Vol. 23, pp. 261-266, 1986.

[3] H. E. Bassil, J. H. Dripps, "Real Time Processing and Analysis of Fetal Phonocardiographic Sensor", *Clin. Phys. Physiol. Meas.*, Vol. 10, Suppl. B, pp. 67-74, 1989.

[4] H. Cao, D.E. Lake, J.E. Ferguson, C.A. Chisholm, M.P. Griffin, J.R. Moorman, "Toward Quantitative Fetal Herat Rate Monitoring", *IEEE Trans. Biomed. Eng.*, Vol. 53, No. 1, pp.111-117, Jan. 2006

[5] P.A. Warrick, D. Precup, E.F. Hamilton, R.E. Kearney, "Fetal Heart Rate Deceleration Detection from the Discrete Cosine Transform Spectrum", *Proc. IEEE Eng. in Med. and Biol. 27th Ann. Conf.*, Shanghai, China, Sep. 1-4, 2005

[6] F. Kovács, M. Török, I. Hábermajer, "An Improved Phonocardiographic Method for Long-Term Fetal Heart Rate Monitoring", *IEEE Trans. Biomed. Eng.,*Vol.47, No.1, pp. 225-230, Jan. 2000.

[7] K. Maeda et al., "Neural Nerwork and Expert System in the Analysis of Intrapartum Fetal Heart Rate", *Journal of Maternal-Fetal Investigation*, vol.8, No.4, pp. 163-171, Dec. 1998.

[8] M. Török, F. Kovács et al., "Method and Apparatus for Measuring Fetal Heart Rate and an Electroacoustic Sensor for Receiving of Fetal Heart Sounds", *US. Patent* No. 6.245.025, June 12, 2001.

[9] F. A. Manning et al, "Fetal Biophysical Profile: A Prospective Study in 1184 High-Risk Patients", *Am. J. Obstet. Gynecol.*, Vol. 140, pp. 507-514, Nov. 1978

[10] M. Akay, H. Szeto, "Analyzing Fetal Breathing Rates Using Matching Pursuits", *IEEE Eng. in Medicine and Biology*, pp.195-198, March/April 1995.

[11] H. G. Goovaerts, O. Rompelman, H. P. van Geijn, "A Transducer for Detection of Fetal Breathing Movements", *IEEE Trans. Biomed. Eng.*, Vol. 36, No. 4, pp. 471-478, Apr. 1989.

[12] M. Signorini et al., "Linear and Nonlinear Parameters for the Analysis of Fetal Heart Rate Signal from Cardiotocographic Recordings", *IEEE Trans. Biomed. Eng.*, Vol. 50, No. 3, pp. 365-374, March 2003.

[13] J. Patrick, R. Natale, B. Richardson, "Patterns of Human Fetal Breathing Activity at 34 to 35 Week's Gestational Age", *Amer. J. Obstet. Gynecol.*, Vol. 132, pp. 507-513, 1978.

[14] J. C. Dornan et al., "The Rate and Regularity of Breathing Movements in the Normal and Growth-Retarded Fetus", *Brit. J. Obstet. Gynecol.*, Vol. 91, pp. 31-36, 1984.

[15] L-G. Durand, P. Pibarot, "Extraction of the Aortic and Pulmonary Components of the Second Heart Sound using Nonlinear Transient Chirp Signal Model", *IEEE Trans. Biomed. Eng.,* Vol.48, No.3, pp. 277-283, March 2001.