Rule Based Signal Processing for Ambulatory Fetal Monitoring

Mohd Alauddin Mohd Ali, Muhammad Ibn Ibrahimy, and Edmond Zahedi

***Abstract*—An algorithm has been developed for the simultaneous measurement of the fetal and maternal heart rates from the maternal abdominal electrocardiogram during pregnancy and labor. The algorithm is based on digital filtering, adaptive thresholding, statistical properties in the time domain and differencing of local maxima and minima. The technique has been developed through a combination and modification of earlier techniques making it suitable for ambulatory monitoring. Eighteen volunteers at various weeks of gestation were studied for the fetal heart rate detection. The computation complexity is such that the developed algorithm can extract both maternal and fetal heart rates in real-time utilizing a single-lead configuration.**

***Index Terms*—Fetal heart rate, digital filtering, abdominal ECG.**

Glossary of Terms

AECG	Abdominal Electrocardiogram
BPM	Beats Per Minute
EMG	Electromyogram
FECG	Fetal Electrocardiogram
FHR	Fetal Heart Rate
MECG	Maternal Electrocardiogram
MHR	Maternal Heart Rate
SNR	Signal-to-Noise Ratio

I. Introduction

Fetal heart rate (FHR) variations observed during pregnancy and labor have commonly been used as indirect indications of the fetal condition. A baseline FHR between 120 and 160 beats per minute (BPM) with a 5 to 25 BPM variability is considered normal, while a baseline rate below 120 BPM (bradycardia) or above 160 BPM (tachycardia) may hint to health hazards [1]. Possible etiologic factors of tachycardia are fetal hypoxia, maternal fever, parasympathetic and sympathetic drugs. Bradycardia of lesser than 120 BPM suggests fetal distress. FHR acceleration and deceleration (rise and fall of the FHR respectively) occur normally for less than 10 minutes either in connection with uterine contraction (periodically) or independently (sporadically). The shapes of these changes in the FHR are further classified into uniform or variable [2]. Changes of FHR for a duration of greater than 15 minutes are considered baseline changes. Baseline rate is elevated with maternal anxiety, maternal fever or immaturity of the fetus. Most women give birth between 38 and 42 gestational weeks of pregnancy, with an average of 40 weeks. FHR monitoring performed weekly from 34 weeks of gestation may identify the need for urgent delivery [3]. Generally, a FHR recording limited to twenty minutes is taken at the clinic as representative of the heart rate variation [4]. FHR abnormalities may however appear outside this interval and may therefore not be detected during short-term monitoring. The ability to perform long-term (e.g. 24 hours) monitoring of the FHR would thus provide more information on the fetal condition [5].

Doppler ultrasound has become a standard procedure for monitoring the FHR abdominally in the clinic, but attempts to produce a portable system have not been successful because of its sensitivity to movements. The expectant mother needs to be in the recumbent position and limit her physical activity during ultrasound FHR monitoring. In addition, changes in the position or orientation of the transducer with respect to the fetus will affect the signal [6], rendering this technique unsuitable for long-term FHR monitoring. Although there is no significant evidence from clinical data that short-term exposure to ultrasound at this low-power level is harmful to the fetus, complete safety of long-term exposure has yet to be established.

Methods utilizing the abdominal electrocardiogram (AECG) seem to be better candidates for long-term monitoring [7]. The major advantages of these methods are them being non-invasive and not exposing the fetus to external energy while allowing instantaneous FHR recording. As these techniques are heavily based on signal processing, advances in microelectronic technology have paved the way towards miniaturization and low-power consumption, thus portability. The main difficulties encountered in determining the FHR from the AECG signal are the interference due to the maternal electrocardiogram (MECG), electromyogram (EMG) and motion artifact. Obtaining a fetal ECG (FECG) with a favorable signal-to-noise ratio (SNR) is dependent on the gestational age of pregnancy. It has been reported that while during 20 to 26 weeks the amplitude of the abdominal FECG remains somewhat constant, during the period of 26 to 34 weeks of gestation it is significantly attenuated [8], due to the covering of the fetus with vernix, which is a poor conductor of electricity. From 34 weeks to delivery the amplitudes vary depending on the electrode location on the abdomen [9]. To overcome the above problems, Khamene and Negahdaripour [10] use the thoracic MECG to cancel the abdominal MECG. These methods require more than

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three leads, which is inconvenient for the patient during long-term monitoring. A single-lead method developed by Mohd Ali [11] uses an average MECG complex formed from previously detected MECG complexes of the AECG signal using peak detection or correlation to cancel the actual MECG complex. As MECG complexes can have different amplitudes and durations, it is not possible to obtain a perfect match between the average MECG complex and the actual MECG complex in all cases. Consequently, MECG subtraction residues lead to errors by producing false fetal peaks. Another single-lead method proposed by Akselrod et al. [12] utilizes the difference in the frequency spectra of the maternal and fetal complexes to separate the maternal signal from the AECG.

In the present study, we have developed an algorithm utilizing the AECG to determine the fetal heart rate. Only a single abdominal lead signal has been utilized to extract the FHR, while attention has been paid so that the algorithm could be implemented in real-time using a microprocessor-based system. To this effect, computations have been kept low in volume and complexity.

II. METHODS

A. Pre-detection Filtering

The developed algorithm processes a single-lead maternal AECG sampled at 500 Hz with 12-bit resolution. Abdominal ECG signals are of relatively low frequency with most of the important information being contained below 100 Hz. A sampling rate of 500 Hz is sufficient for the observation of the variation in the baseline heart rate [11]. The major routines of the algorithm are shown in Figure 1. Two almost similar sets of operations are used to enhance and then detect the maternal and fetal QRS complexes respectively.

The AECG is first passed through a finite impulse response bandpass filter to reduce power line interference, baseline drift, muscle noise (EMG) and motion artifacts. The highest power density of the R wave of an adult lies between 10 and 30 Hz and that of a fetus between 15 and 40 Hz [13]. Hence the coefficients of the filter are selected such that the lower and upper cut-off frequencies are 10 and 40 Hz respectively. Using a Hamming window, a filter with $N+1$ coefficients, $N=10$ has been empirically found to be effective. The coefficients of the Hamming window used are:

$$w_k = 0.54 - 0.46 \cos\left(\frac{2\pi k}{N}\right), \qquad k = 0, \ldots, N \tag{1}$$

B. Maternal QRS Detection

The detection of maternal QRS complexes begins with cross-correlating the filtered signal with an average maternal QRS template. The initial template is a crude approximation of the signal after being filtered but is sufficient in most cases before actual signal average is used. The cross-correlation (the matched filter) output of the signal x at each instant n with the template $s(k)$ with $M+1$ equally spaced points is given by:

$$y(n) = \sum_{k=0}^{M} h(k)x(n-k) \tag{2}$$

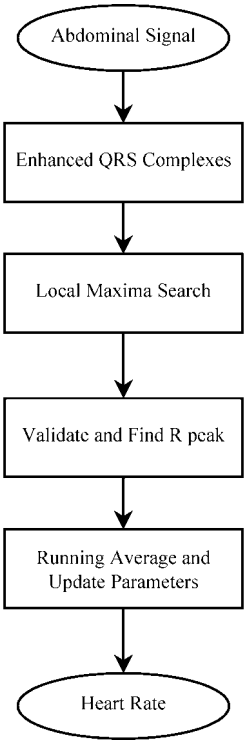


Fig. 1. Flow diagram of the algorithm.

where

$$h(k) = \begin{cases} s(M-k) & 0 \le k \le M \\ 0 & \text{elsewhere} \end{cases}$$

The template $s(k)$ over 80 ms has been empirically found to be optimized for the detection of maternal QRS complexes when $M=8$. The width of the template is based on the normal width of the maternal QRS complex [14]. $s(k)$ is continuously updated with the detection of new R peaks to take into consideration the variation of shape of the maternal QRS complexes in the AECG.

The local maxima search routine measures the slope of the cross-correlated output by:

$$y'(n) = y(n) - y(n-1) \tag{3}$$

and assumes a maximum at sample $n-1$ when the slope changes from $y'(n) \ge 0$ to $y'(n) < 0$. If no maximum is found in the subsequent 20 ms (assumed to be the minimum fetal QRS duration [11]), the sample value $y(n-1)$ and corresponding instant are saved as the local maximum. This 20 ms search interval is necessary to avoid taking small spikes on the slopes of the QRS complexes as maxima.

Three values, $V_{M1} > V_{M2} > V_{M3}$, and their corresponding time instants (largest three local maxima) are stored within an R wave search interval. The length of the search interval is initially 1024 ms (closest value to one second allowing for computational simplicity) and it is then continuously updated after the first RR interval measurement. This search interval duration and the saving of three local maxima assume that the maternal heart rate (MHR) does not exceed 120 BPM which means at most two maternal R peaks can be found in the initial search interval. If only V_{M1} is validated as the R peak then the value V_{M2} is taken as noise. V_{M3} is taken as noise when V_{M2} is also validated as an R peak. The threshold used in the detection is set

initially by assuming a minimum maternal R peak of 10 μV [11] and it is continuously updated based on the levels of both R peak and noise. A possible maternal R peak is assumed to be found when the value V_{M1} exceeds this threshold. V_{M2} is also considered as an R peak if the value is comparable to that of V_{M1} and the resulting heart rate is below 120 BPM, as earlier assumed. Hence the set of criteria:

$$2V_{M2} > V_{M1} \tag{4a}$$

and

$$|t_{M2} - t_{M1}| > 512 \text{ ms} \tag{4b}$$

If V_{M2} exceeds the threshold without the criterion in (4b), the QRS template is compared with the complexes associated with both V_{M1} and V_{M2}. The one with the least mean square error is taken to be the R peak. The other peak is assumed to be a large spike in the signal and its position is saved for use in the fetal R peak validation routine. If V_{M2} has the larger error, its position is saved only if inequality in (4a) applies, because smaller V_{M2} may be associated with an actual fetal R peak.

A running average is performed to average the QRS templates, RR intervals, levels of R peaks and noise. The b-th value of the running average $A(b)$ is given by a weighting of the previous average $A(b-1)$ plus that of the new value $C(b)$ as shown in the following equation:

$$A(b) = [1 - k(b)]A(b-1) + k(b)C(B) \tag{5}$$

where

$$k(b) = \begin{cases} 1/b & b \le B \\ 1/B & b > B \end{cases}.$$

The running averages of noise and R peaks (A_N and A_R) are estimated over B recent values (where $B = 8$ has been empirically found to be effective). Based on these averages, two thresholds, TM_1 and TM_2 are used in the R wave search. They are given by:

$$TM_1 = A_N + \frac{A_R - A_N}{4} \tag{6a}$$

$$TM_2 = \frac{TM_1}{2}. \tag{6b}$$

The adaptation of the threshold to varying R peak and noise levels, and the R wave search interval are based on the method proposed in [15].

If the maximum search limit is reached while the local maximum V_{M1} has a value less than TM_1, then V_{M1} is taken as a possible R peak if it exceeds the second threshold, TM_2. If no such V_{M1} is found, a signal loss is assumed. The local maxima values are then set to zero for the subsequent R wave search. Four latest maternal RR intervals are maintained in record for the purpose of checking coincidences of the maternal with the fetal R waves.

C. Fetal QRS Detection

The MECG complex is ensemble averaged upon detection of a maternal QRS to reduce the maternal contribution from the abdominal signal. This complex is of fixed duration, 160 ms before and 320 ms after the maternal R peak instant. This duration assumes that the average MHR is less than 125 BPM and it should normally include the P and T waves, if any. The resultant waveform of the ensemble averaging at each instant n over the MECG complex is given by:

$$z(n) = \frac{1}{L}\sum_{i=0}^{L-1} y(n-i) \tag{7}$$

where $y(n)$ is the cross-correlation output in (2) and L is the number of samples averaged ($L = 3$ has been empirically found to be effective). The signal is then passed through another bandpass filter with a Hamming window to enhance the fetal QRS complexes. For enhancing the FECG, the coefficients of the filter are selected such that the lower and the upper cut-off frequencies are given by 30 Hz and 40 Hz respectively with $N+1$ coefficients ($N = 40$ is used in (1)). Filter parameters were determined by using the Acqknowledge-III (BIOPAC System, Inc) Biomedical Signal Processing Software for steep cutoff as well as ability for the algorithm to be implemented in real-time by commercially available microcontrollers.

The detection of the fetal QRS complex begins with differencing of local maxima and minima [16] on the output of the second digital bandpass filter when the time marker count, which was initiated at the second accepted maternal R peak, has reached 2048 ms. This duration ensures that the two seconds delayed samples are already within the MECG averaged region of the signal. Observing the waveforms, it is possible to differentiate between fetal events and noise even if the amplitudes are similar. This is partly because of the rapid and large deflections between a local maximum and the following local minimum when a fetal beat occurred. From (3), a minimum is assumed at sample $n-1$ when the slope changes from $y'(n-1) < 0$ to $y'(n) \ge 0$. The absolute value of the difference between successive peak and valley is computed for each max-to-min interval.

The local maxima search routine is performed on the output of the differencing of local maxima and minima routine, and three largest maxima, $V_{F1} > V_{F2} > V_{F3}$ are kept as before. The initial search interval is 640 ms so that at most two fetal R peaks can be found by assuming the FHR does not exceed 187 BPM during the initial search interval. The first search is repeated for another subsequent 640 ms if the largest local maximum, V_{F1} is concurrent as defined in (10) with a maternal QRS complex and V_{F2} is smaller than a threshold or is also concurrent. The threshold used in the FHR detection is set initially by assuming a minimum fetal R peak of 5 μV [11] and it is continuously updated. The routine is similar to that for the maternal case but uses the following criteria to accept V_{F2} as a possible fetal R peak:

$$1.5V_{F2} > V_{F1} \tag{8a}$$

and

$$|t_{F2} - t_{F1}| > 320 \text{ ms} \tag{8b}$$

The second search is repeated if the accepted first fetal R peak is found to be concurrent with a maternal QRS complex or if

$$2V_{F3} > V_{F1} \tag{9}$$

i.e. the signal is noisy with all its three local maxima having comparable values.

The fetal and maternal QRS complexes are concurrent if

$$|t_F - t_M| < 64\,\text{ms} \tag{10}$$

where t_F and t_M are the fetal and maternal R peak instants, respectively. The range in (10) accounts for possible overlap of the two complexes, which are assumed to have widths of 50 and 80 ms, respectively. The overlap is checked by relating the fetal R peak instant to the four latest maternal RR intervals.

The subsequent fetal R wave detection procedure is the same as that for the maternal R wave using two thresholds, TF_1 and TF_2 which are set as in (6a) and (6b), according to the running average of the R peaks and noise with $B = 8$ in (5). The determination of the fetal R wave search interval is also based on the method proposed in [15]. The second threshold, TF_2 is used when the maximum search limit is reached. A signal loss is assumed when no maximum exceeding the threshold is found. When the second threshold is used to identify a fetal R peak, the peaks are averaged with $B = 4$ so that the first threshold will quickly adapt to the smaller signal.

After a possible fetal R wave is found, a continuation of the search for up to 220 ms is carried out unless the maximum search limit is reached. This forward searching reduces the possibility of false R wave detection with the assumption that the heart rate does not exceed 270 BPM. Then the program branches to the validate and update routines. The validate routine first checks if

$$V_{F3} > TF \tag{11a}$$

and

$$1.5V_{F3} > V_{F1} \tag{11b}$$

where TF is the threshold used to detect V_{F1}. These conditions mean that the fetal R peak was obtained in a very noisy signal. Otherwise, similar checks are made with V_{F2}, where

$$V_{F2} > TF \tag{12a}$$

and

$$1.5V_{F2} > V_{F1} \tag{12b}$$

also imply a noisy signal.

If V_{F1} is the only maximum above the threshold then it is taken as a fetal R wave. If V_{F2} also exceeds the threshold, then V_{F1} is checked for coincidence with possible spikes by relating its instant to the four maternal values which are kept in the record. The spike position, t_S and the position, t_{F1} in the signal associated with the local maximum, are compared for

$$|t_S - t_{F1}| < 40\,\text{ms} \tag{13}$$

which allows for the difference in correlation delay when obtaining t_S and t_{F1}, respectively. If V_{F1} is identified as a large spike in the signal, then V_{F2} and V_{F3} are assumed to be the fetal R peak and the noise, respectively.

Thresholds and search interval limits are updated according to the procedure described earlier and the local maxima values are then set to zero for the subsequent R wave search.

D. Experimentation

To ascertain the effectiveness of the algorithm, AECG signals from the abdomen of eighteen volunteer pregnant women carrying single fetus at gestational ages between 21 and 41 weeks were recorded. Subjects were either in the supine position, sitting upright or sitting reclined throughout the recording. Five subjects were allowed to exert some movements in order to check the effectiveness of the algorithm during movement (ambulatory mode). Two of the recordings were done when subjects were standing up and walking as with the use of long-term monitoring, these women can maintain daily normal activities and work. All recordings were performed with the vertical electrode placement (epigastrium and infra-umbilical), the third electrode being located on the right wrist. The algorithm was tested using the recorded AECG data under different signal/heart rate conditions, namely a relatively noiseless (clean), noisy, monotonic and containing FHR acceleration. The performance of the heart rate detection is assessed using the following formula [16]:

$$P = 100 \times \frac{N_R - N_u - N_e}{N_R} \tag{14}$$

where

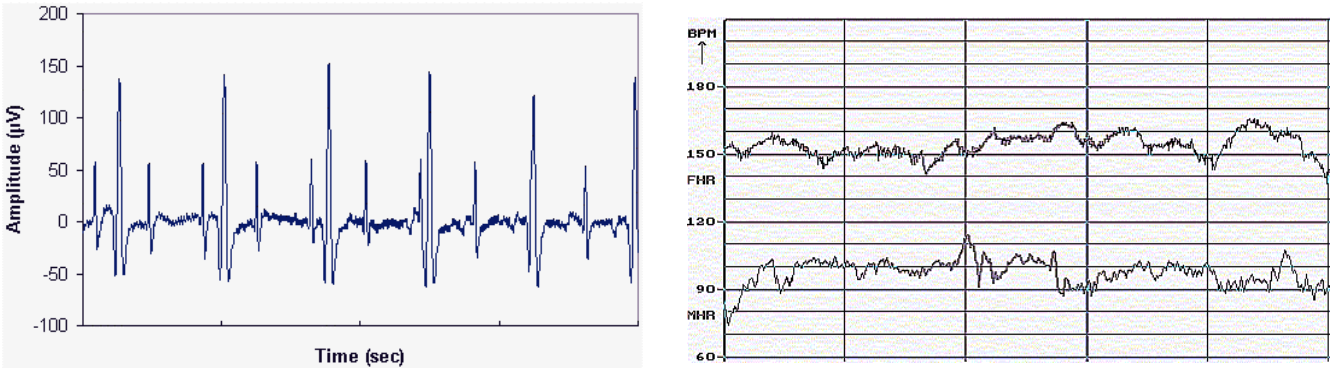
N_R = number of actual R peaks in the waveform
N_u = number of undetected R peaks
N_e = number of extraneous peaks

This measure is based on *a-priori* knowledge of the locations of all R peaks in a specific waveform. In our study this knowledge is obtained via visual inspection of the ECG trace.

To facilitate the identification of the gestation age, each record is represented by a letter from the alphabet followed by the gestation age in weeks.

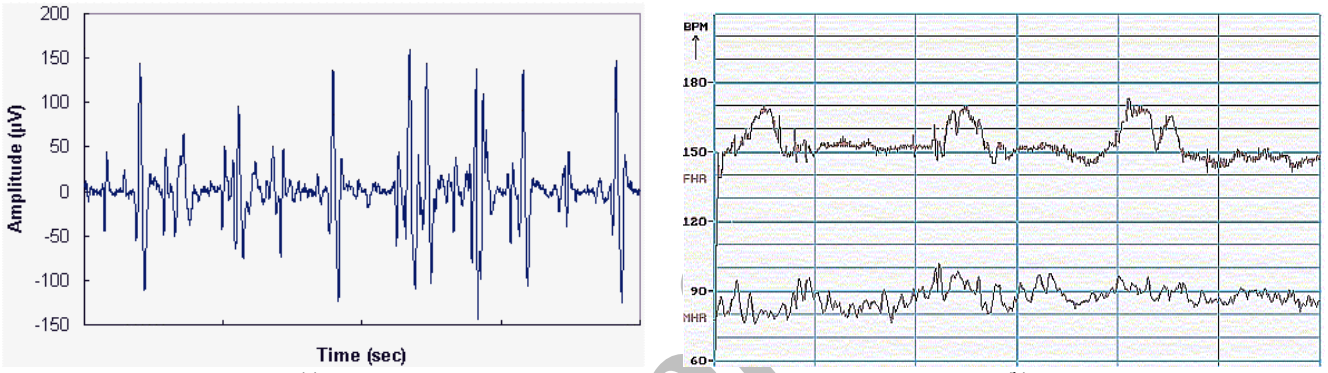
III. RESULTS

An example of AECG with very low interfering noise and motion artifact from subject C39 is shown in Fig. 2(a). Corresponding traces of FHR and MHR obtained by the proposed algorithm are displayed in Fig. 2(b). Sample traces of AECG, FHR and MHR obtained by the algorithm are presented in Figs. 3(a) and 3(b) where subject J37 was in the sitting position throughout the recording. Fig. 4(a) shows an AECG with intermittent maternal movement during recording from subject L38 while corresponding heart rate traces with acceleration in the FHR are shown in Fig. 4(b). Fig. 5(a) shows an AECG with less visibility of fetal peaks from subject N38 and corresponding traces obtained by the proposed algorithm are displayed in Fig. 5(b). A number of spikes and noise episodes due to considerable motion artifact are visible in Fig. 6(a) from subject Q41. When the heart rate could not be detected due to either much EMG noise during contraction or motion artifact, traces are left blank as in one segment in the last 90 seconds in Fig. 6(b). The FHR and MHR detection performances of the algorithm are listed in Table I (mother remaining still) and Table II (mother allowed to exert some movement). Patient's posture and coincidences of the maternal with the fetal complexes are also included in these tables. The percentages of occurrence of noisy/spikes detected are given in both tables. A comparison between



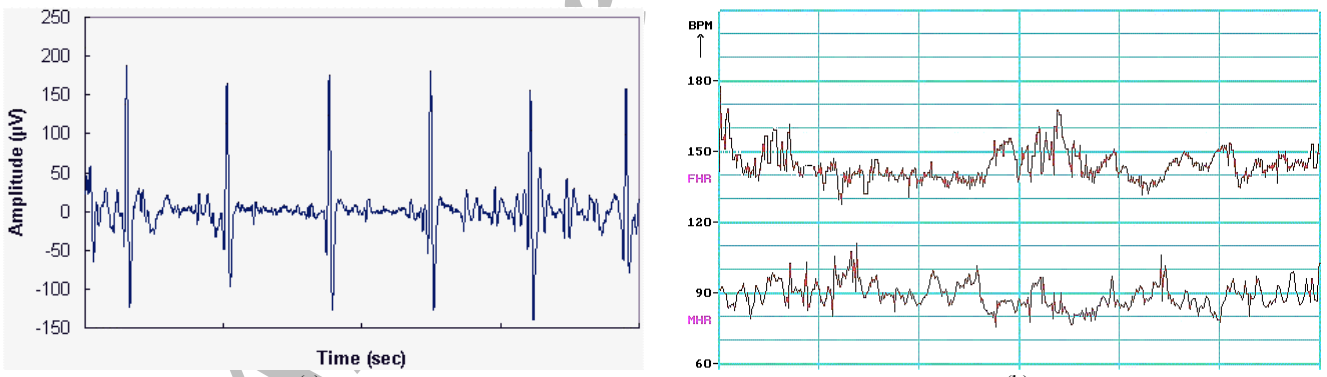
(a) (b)

Fig. 2. (a) Example of relatively noiseless AECG from subject C39 (each vertical division = 1 sec.), (b) FHR (upper) and MHR (lower) traces from subject C39 using the proposed algorithm (each vertical division = 90 sec.).



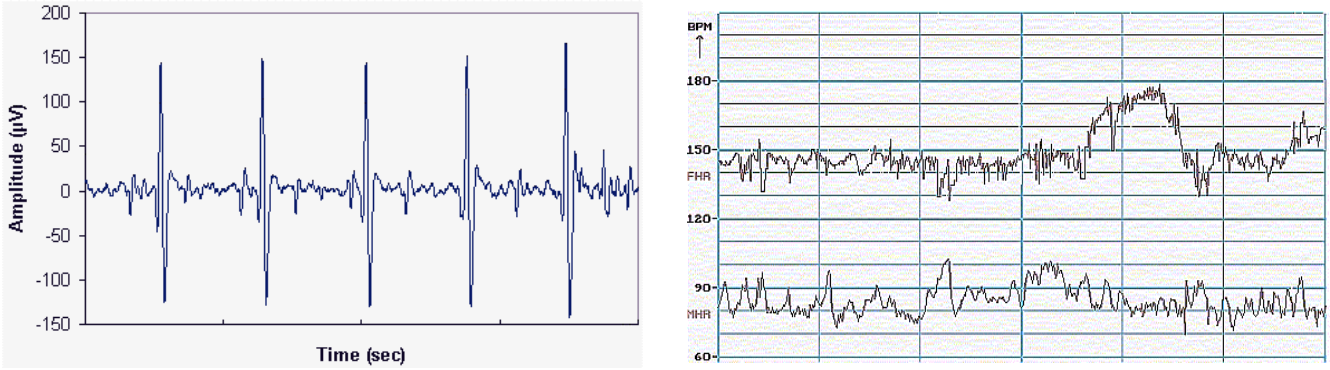
(a) (b)

Fig. 3. (a) Example of AECG from subject J37 at sitting position (each vertical division = 1 sec.), (b) . FHR (upper) and MHR (lower) traces from subject J37 using the proposed algorithm (each vertical division = 90 sec.).



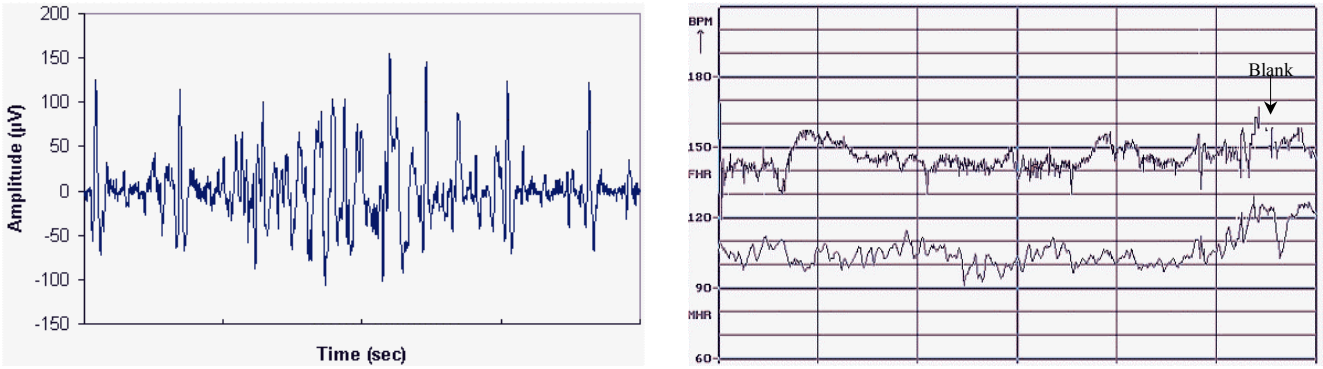
(a) (b)

Fig. 4. (a) Example of AECG from subject L38 with noise due to maternal movement (each vertical division = 1 sec.), (b) FHR (upper) and MHR (lower) traces from subject L38 using the proposed algorithm (each vertical division = 90 sec.).



(a) (b)

Fig. 5. (a) Example of AECG from subject N38 with a low FECG-to-noise ratio (each vertical division = 1 sec.), (b) FHR (upper) and MHR (lower) traces from subject N38 using the proposed algorithm (each vertical division = 90 sec.).



(a) (b)

Fig. 6. (a) Example of AECG from subject Q41 with considerable maternal movement (walking) (each vertical division = 1 sec.), (b) FHR (upper) and MHR (lower) traces from subject Q41 using the proposed method.

TABLE I
PERFORMANCE OF THE PROPOSED ALGORITHM EVALUATED BY A-PRIORI KNOWLEDGE OF R PEAKS. ALL SUBJECTS IN REST POSITION

Subject	Week of Gestation	Patient's Posture During Recording	Signal Condition Counts (%) (During FHR Detection)			Detection Performances (%)	
			Coincidence [Eq. (10)]	Noisy [Eqs. (11) - (12)}	Spike [Eq. (13)]	MHR	FHR
A40	40	Supine position	23.3	14.7	0.2	99.3	96.4
B39	39	Supine position	27.6	17.3	0.2	99.0	93.8
C39	39	Supine position	21.0	13.4	0.1	100.0	98.5
D39	39	Supine position	22.9	15.9	0.3	98.1	89.3
E38	38	Supine position	24.8	16.7	0.2	99.1	87.9
F38	38	Supine position	18.3	25.9	0.8	98.2	80.6
G37	37	Supine position	28.1	18.9	0.7	98.9	69.6
H35	35	Supine position	29.0	26.5	0.7	98.6	68.4
I38	38	Sitting	26.8	22.2	0.4	98.8	81.4
J37	37	Sitting	19.2	16.4	0.3	99.7	88.5
K31	31	Sitting	48.8	39.6	2.0	99.7	32.8

TABLE II
PERFORMANCE OF THE PROPOSED ALGORITHM EVALUATED BY A-PRIORI KNOWLEDGE OF R PEAKS. SUBJECTS ALLOWED SOME MOVEMENTS

Subject	Week of Gestation	Patient's Posture During Recording	Signal Condition Counts (%) (During FHR Detection)			Detection Performances (%)	
			Coincidence [Eq. (10)]	Noisy [Eqs. (11) - (12)}	Spike [Eq. (13)]	MHR	FHR
L38	38	Supine position / movement	18.8	20.1	1.1	99.4	83.5
M38	38	Supine position / movement	23.2	19.6	0.8	99.6	85.8
N38	38	Supine position / movement	26.6	27.4	2.5	98.7	68.5
O36	36	Supine position / movement	24.1	36.2	5.1	96.3	62.2
P21	21	Supine position / movement	26.3	49.4	2.4	95.7	26.6
Q41	41	Supine position / walk	24.3	18.7	2.1	98.1	87.6
R34	34	Supine position / walk	41.6	62.4	7.1	91.2	52.5

average FHR detection performances (listed in Table III) of a previous method developed by Mohd Ali [11] and the proposed method is displayed in Fig. 7.

IV. DISCUSSION

A. Maternal Heart Rate

Tables I and II show that a very good (>98%) performance as defined by (14) has been obtained for the measurement of the MHR in most cases. The higher performance listed in Table I is due to the fact that subjects in these cases were all at rest. However, slightly lower performances for D39, F38, G37, H35 and I38 are attributed to the presence of large episodes of noise. In Table II (where subjects were allowed to exert some movements), the performance for subject R34 (for example) is degraded (91.2%) due to extensive maternal movement as the subject was asked to walk. Performances in subjects O36 and P21 were also relatively low (96.3% and 95.7%, respectively) as the maternal ECG signals were of lower amplitude in these

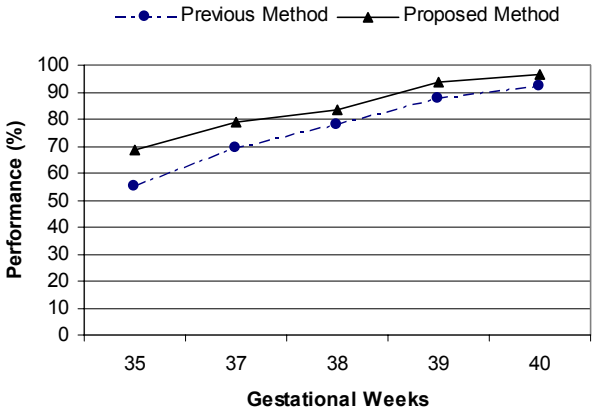


Fig. 7. FHR detection performances as defined in (14) versus gestational weeks (data from Table III). No data was available for 36 weeks of pregnancy.

cases, besides the existence of motion artifact. Motion artifact has less affected the detection performance for subjects N38 and Q41 with a performance of 98.7% and 98.1% ,respectively.

B. Fetal Heart Rate

As expected, a similar level of performance cannot be achieved with the fetal heart rate due to the lower signal-to-noise ratio with small amplitude fetal complexes compounded by motion artifacts in some cases. Signals at lower gestational weeks have produced very poor FHR traces in subjects K31, O36, P21 and R34. Fig. 7 shows that the performance obtained by the proposed method is better than that of the previous method. It also shows that the obtained performance seems to increase regularly with the gestational age before reaching a plateau after 39 weeks. Generally speaking it can be seen from data in Tables I and II that when the sum of the percentages of signal condition with the "noisy" and "spike" levels exceeds 30%, very poor FHR traces are produced. Obtained performances from the abdominal signals with a very small fetal R peak in subjects K31 and P21 are not acceptable (lower than 50%). Good performance values (>80%) for subjects A40, B39, C39, D39, E38, F38, I38, J37, L38, M38 and Q41 are due to the consistent presence of large fetal signals at higher gestational weeks. Performances in subjects G37 and H35 were relatively poor (69.6% and 68.4% respectively) due to some sections of the signal having high noise levels combined with small fetal complexes. In record N38, abdominal signals with a low FECG-to-noise ratio plus motion artifacts produced comparatively poor (68.5%) FHR trace.

In cases where signals are affected by noise and motion artifacts, errors are encountered in the FHR determination. However, correct FHR detection is resumed when movements no longer affect the signals. When the fetal signal or its ratio to that of the mother's is small, consistent detection of the fetal R peak is more difficult and the performance values for the resulting trace decrease. This happens especially when the gestational age is lower. Episodes of FHR acceleration due to fetal movements as expected in normal pregnancy are clearly visible on the traces obtained by the algorithm. Observing MHR and FHR detection performances for subject C39 in Table I, it can be seen that the proposed algorithm can in this case separate the fetal and maternal R waves from the composite AECG even where they coincide for slightly more than one-fifth of the time (21%).

TABLE III
PERFORMANCE COMPARISON OF A PREVIOUS METHOD [11] WITH THE PROPOSED METHOD

Week of Gestation	Average Detection Performances of a Previous Method (%)		Average Detection Performances of the Proposed Method (%)	
	MHR	FHR	MHR	FHR
40	99.1	92.0	99.3	96.4
39	98.9	87.6	99.0	93.9
38	98.1	77.5	98.7	83.3
37	98.8	69.0	99.3	79.0
35	96.5	54.8	98.6	68.4

C. Suitability for Ambulatory Real-Time Implementation

The number of mathematical operations used in the algorithm are 108 multiplications, 133 divisions, 245 additions, 182 subtractions and 135 comparisons. Currently available low-power microcontrollers are capable of addition, subtraction, shift, logical operation and multiplication in a single cycle instruction. For example, the instruction cycle for the PIC17C44 from Microchip Technology Inc. is only 120 ns for a 33 MHz clock allowing for more than 16,600 instructions to be executed between each sample, real-time operation can thus be guaranteed.

V. CONCLUSION

The performance achieved for the heart rate measurements from the AECG show that the proposed algorithm can extract both maternal and fetal heart rates utilizing a single-lead configuration. The single-lead feature is desirable from the comfort point of view of the patient especially when the duration of monitoring is long. The algorithm is also found to be effective in resolving the FHR when fetal and maternal R waves are overlapped.

Improvements to the R peak detection capability of the algorithm are still achievable with enhanced procedures such as the normalization of the cross-correlation outputs and variable MECG complex duration (to take into account for cases when the MHR exceeds 125 BPM). The length of the FIR filter was chosen by taking into account the ability of the algorithm to be implemented in real-time on commercially available microcontrollers. An increased number of coefficients with a suitable processor can offer a steeper cutoff, thus a better FECG enhancement. The sensitivity of the algorithm to motion artifacts and muscle noise may also be reduced with the incorporation of more rules in its RR interval validation schemes.

As expected, a favorable FECG's signal-to-noise ratio (visually detectable FECG) is a definitive enhancement in future research. This will allow a reliable FHR determination at a lower gestational age (below 35 weeks). Implementation of the algorithm on a microprocessor-based system will provide the clinician with real-time FHR/MHR monitoring. This tool will be useful in the assessment of the fetal condition and its relationship to that of the mother's.

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