IJIAP International Journal of Information Analysis and Processing, Vol. 2, No. 1, April 2009, pp. 1-6

Independent Component Analysis for Fetal Heart Rate Detection Using Photoplethysmography

Malek M. A. Harbawi, Othman O. Khalifa and M. A. Mohd Ali

Abstract: Photoplethysmography (PPG) is an optoelectronic technique for measuring and recording changes in the volume of body parts. These changes are associated with each heart beat and acquired by pulse oximetry. Fetal heart rate (FHR) monitoring using PPG is a challenging task since the acquired signals present the pattern of both fetal and maternal hearts. The effect of maternal component, noise and artifacts on fetal component makes the separation of FHR very difficult. In this paper, we study the applicability of independent component analysis (ICA) to FHR detection using PPG. The study was conducted using emulated signals to mimic the pulsation nature of both maternal and fetal hearts. The outcome of this experiment shows encouraging results in terms of the extraction ability of ICA, which can perform well even when fetal-to-maternal signal-to-noise ratio ($S_F N_M R$) drops to -276 dB.

Index Terms: PPG, pulse oximetry, FHR monitoring, ICA, SNR.

I. INTRODUCTION

Pulse oximetry is a non-invasive device for measuring and recording the changes in the volume of body parts such as fingers, earlobes, toes and etc. caused by the changes in the volume of arterial oxygenated blood, associated with cardiac contraction [1]. Generally, pulse oximetry probe consists of two LEDs and one photodetector; and it can be used for estimating arterial blood oxygenation, venous blood oxygenation and heart and respiratory rates [1], [5], [6]. Moreover, pulse oximetry acquires the photopletysmographic (PPG) signal which has a significant contribution in the non-invasive medical diagnoses [2], [3]. PPG signal is based on the absorption properties of vascular tissue when it is trans-illuminated by light. This light is shone by pulse oximetry probe, placed on a certain extremity in two different ranges, namely red (660nm) and infrared (940nm) consecutively. The emitted light beams are made to traverse extremity tissues. During tissues penetration, light beams will be affected by reflection, absorption and scattering in the tissues and blood. The emerged light, which is modulated by both tissues attenuation and pulsation nature of blood flow, is measured using a suitable photodetector. Depending on the extremity properties, photodetector can be placed either opposite (transmitted mode PPG) or beside (reflected mode PPG) the LEDs. The waveform in the received light allows the absorbance effect of the arterial blood to be identified from those of non-pulsatile venous blood and other tissues. The variations in the photodetector current are assumed to be related to the blood volume changes underneath the probe, which are electronically amplified and recorded as a voltage signal [2].

Malek M.A. Harbawi & Othman O. Khalifa, is Department of Electrical Engineering, International Islamic University, 50728, Kuala Lumpur, Malaysia, *E-mail: malyk19@yahoo.com*

M. A. Mohd Ali, is Department of Electrical, Electronic and System Engineering, Faculty of Engineering, University Kebangsaan Malaysia, 43600 UKM, Selangor, Malaysia

Over the past two decades, there has been a renewed interest in PPG monitoring techniques; its applications have been widely employed in the biomedical field. Its usage ranges from the battlefield to the sport-field as a monitoring device [16], [17]. It gains its advantages due to its non-invasivity, versatility, portability, rapid response, real-time measurement ability and low-cost components comparing to other medical devices, such as ECG sensors. In this regard, PPG monitoring has been suggested as a monitoring technique for fetal heart rate (FHR) [7-15]. In obstetrics, FHR monitoring is a routine procedure performed at hospital for assuring fetal well-being [18]. Parameters extracted from FHR assist in the detection of potential hazardous such as fetal hypoxia, acidemia and distress [14], [18]. The early detection of such diseases increases the effectiveness of the treatment [19]. The most applied FHR monitoring standard in obstetrical wards is Doppler ultrasound which is performed from onset of labor until the loss of amniotic fluid [20]. This technique relies on Doppler signal shift induced by fetal heart contractions. The main limitation of this technique is its sensitivity to the movement caused by fetal and maternal breathing. In addition, the effect of Doppler ultrasound on the fetus is not completely investigated, thus it is not recommended for the long-term monitoring purpose [19].

FHR has also been conducted using fetal electrocardiogram (FECG). This technique has been used invasively and non-invasively. An invasive FECG is applied using a scalp electrode and can only be used during labor but this technique is not free of hazard and needs at least three to four leads, which increase the complexity [14], [19]. Alternatively, non-invasive FECG has been proposed [19], [21], [22]. This technique is normally utilized later in pregnancy due to its low SNR [14]. Beside the previous methods, FHR using optical methods has also been proposed and investigated. Similar to FECG, the early foundation of this field started with the invasive fetal pulse oximetry (FPO) probe. The invasive FPO was conducted by placing the probe through the birth canal onto a part of the fetus. Hence, the applicable time for this scenario is during labor. Moreover, the probe requires a cervical dilation of 2 cm and ruptured membranes to obtain the readings [11]. More recently, the trans-abdominal near-infrared (NIF) spectroscopy of fetal cerebral blood oxygenation has been proposed [7], [8]. These early researches proposed the concept and tested it on a phantom model. An extension to the previous studies, [9] and [10] have obtained FHR by placing a pulse oximetry on the maternal abdomen (MA). The main contribution of these studies was the investigation of the most suitable wavelengths for signal acquiring purpose. According to [9] and [10], the selection of wavelengths in the range of (670-720 nm and 825-925nm) offers the optimal acquired FPO signal and reduces the saturation errors. Furthermore, these studies investigated the source-detector separation issues and suggested that large source-detector separation maintains less optical shunting errors. Moreover, the suggested model has the advantage of the real-time applicability which can be performed for ante-partum and intra-partum continuously [11]. A more complete and precise trans-abdominal FPO system has been developed in [12-15]. This model was employed low power LEDs as light sources to ensure the safety of fetal growth, especially fetal eyes [12]. However, lowering the emitted optical power obviously leads to a lower SNR due to the existence of the ambient light and the huge maternal pulsation comparing to the fetal one. Thus, the main idea was focused on designing an effective filtering technique to extract FHR pattern under a low SNR conditions. An evaluation of an adaptive noise cancellation (ANC) technique to extract a clear interpretation of fetal PPG (FPPG) was conducted. In addition, a simple optical model to simulate the maternal and fetal pulsation nature has been proposed and tested. The results of this model using semi-synthetic signals show that the minimum applicable S_FN_MR is -25 dB [13-15] and an ANC algorithm is able to extract clear fetal peaks even when the S_FN_MR drops to -34.7 dB. Nevertheless, in reality fetal and maternal interaction may cause of very low S_FN_MR which leads to failure in the extraction process. Hence, a lower S_FN_MR model is recommended.

In this paper, an evaluation of ICA separation ability to FHR using trans-abdominal FPO is conducted. The advantage of using ICA is allowing the detection of FHR even when the S_FN_MR is very low. In addition, this algorithm has a fast convergence and easy separation procedure.

II. METHODOLOGY

(A) Data Acquisition

To test ICA applicability to FHR detection, there must be acquired signals. Unfortunately, currently such signals are not yet available. Therefore, to proceed with the experiment, emulated signals were used. The emulated signals were taken from a previous experiment [14] with the following specifications: PPG recording systems (Dolphin Medical, Inc.) were used to acquire the PPG signals at rate of 275 Hz and 16-bit resolution which are specified by the manufacturer. The system consists of two finger probes (Digital Oximetry Sensor 201) and two laptops were used for viewing and saving the PPG signals. The laptops were connected via serial ports and one synchronization circuit is used to apply a train of pulses that will be used for off-line synchronization.

(B) Preprocessing

In this stage, the acquired PPG signals refined by different processes started by removing any linear trend in the signals, then the signals were set to zero mean. To eliminate the undesired low and high frequencies, the signals filtered with a band-pass filter has cut off frequencies 0.6-15 Hz, which rejects the low respiration signals (0.1-0.55 Hz) and the high frequency noise that comes from other sources such as electric light and instrumentation noise [14]. The last preprocessing step is normalizing the amplitudes of acquired PPG signals from -1 to +1.

(C) Signal Emulation

For the emulation purpose, PPG signals were acquired from two different healthy subjects, subject A and subject B (right index fingers), so that there are the independent components. The data acquired from subject A is applied directly to preprocessing steps and then rescaled. This PPG signal corresponds to the maternal PPG (MPPG), whereas the data that is acquired from subject B is down sampled by factor of 2.7 (corresponding to 162 BPM), rescaled and then passed to preprocessing steps as mentioned above. The resulted PPG signal corresponds to the FPPG. The complete preprocessing steps are shown in Fig. 1. In fact, the down sampling plays the role of increasing the frequency of the signal in such a way that fits the fetal heart beats rate where fetal heart beats at a much higher rate than the maternal ones [14]. In order to test the performance of ICA algorithm under different ranges of S_FN_MR, signal scaling stage was set as one of the emulation steps. Scaling parameters will be presented in the following section.

(D) ICA Model

All preprocessing and signal emulation were done in the MATLAB environment with data fed to the ICA toolbox developed by RIKEN Brain Signal Institute, Japan [4]. Algorithm for multiple unknown source extraction (AMUSE) was chosen for achieving the extraction task. To test the applicability of ICA algorithm, first we need to quantify the minimum S_FN_MR for typical maternal and fetal tissues. In this regard, it was shown in [14] that, using a simple optical model for both maternal and fetal tissues, the expected S_FN_MR (the ratio of average fetal PPG power to the maternal one) ≈ -25 dB. On the other hand, according to [11] and with source-detector separation approximately

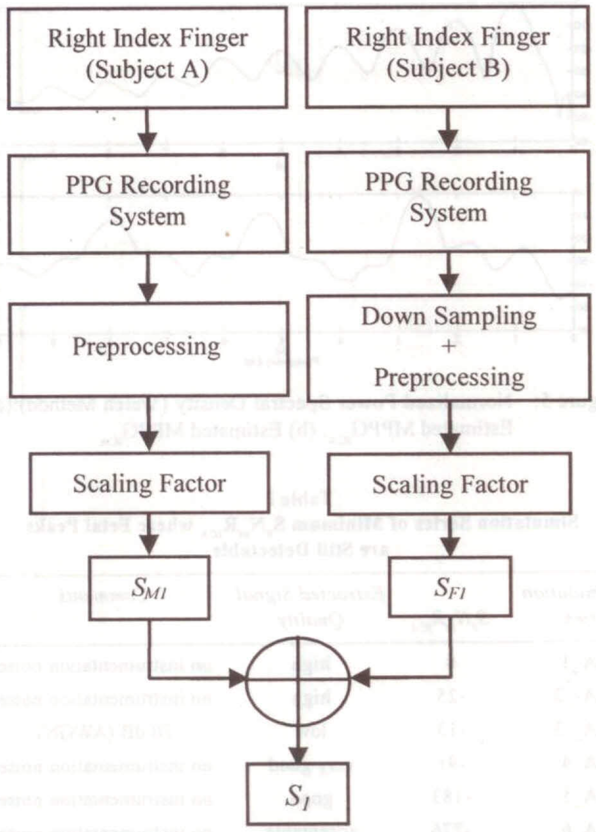

Figure 1: Signal Emulation Steps

twice the depth of the fetus (7-11cm), the expected $S_F N_M R \approx 0$ dB.

Equation 1 represents the ICA model for the evaluation purpose of this study. As we can see from equation 1, to estimate the fetal heart rate ICA algorithm needs at least two probes to extract FPPG signal. Consequently, signal emulation steps, Fig. 1, must be applied twice with different amplitude factors in order to get the required signals.

$$\begin{pmatrix} S_1 \\ S_2 \end{pmatrix} = \begin{pmatrix} a & b \\ c & d \end{pmatrix} \begin{pmatrix} S_M \\ S_F \end{pmatrix} + \begin{pmatrix} g_1 \\ g_2 \end{pmatrix} \quad (1)$$

where

S_1 first emulated maternal abdominal PPG (MAPPG) signal

S_2 second emulated MAPPG signal

S_M an estimated MPPG signal

S_F an estimated FPPG signal

g_1 and g_2 instrumentation noise (assumed as AWGN)

a, b, c and d amplitude controlling factors

However, the most important parameter to be considered here is $S_F N_M R$. This parameter is playing the major role in evaluating ICA algorithm. Since the proposed method must have at least two probes, this leads to two calculated $S_F N_M R$. These parameters are represented by equations 2 and 3. Controlling factors (a, b, c and d) which are presented in equations 1, 2, 3 and 4 were modified empirically in order to test the minimum applicable $S_F N_M R$. Finally, the minimum $S_F N_M R_{ICA}$ is defined by equation 4.

$$S_F N_M R_1 = 10\log\left(\frac{b \times \sum_{i=1}^{end} (S_F(i))^2}{\sum_{i=1}^{end} (a \times S_M(i) + g_1(i))^2}\right) \quad (2)$$

$$S_F N_M R_2 = 10\log\left(\frac{d \times \sum_{i=1}^{end} (S_F(i))^2}{\sum_{i=1}^{end} (c \times S_M(i) + g_2(i))^2}\right) \quad (3)$$

$$S_F N_M R_{ICA} = \max(S_F N_M R_1,\ S_F N_M R_2) \quad (4)$$

III. RESULTS

(A) Applicability Test

As mentioned in the methodology, evaluating ICA in this study is based on typical marginal $S_F N_M R$ value. A series of experiments were conducted started with $S_F N_M R$ = 0dB and $S_F N_M R$ = -25dB. These two points give an impression about ICA applicability in the typical situation [11] and [14], so that further circumstances can be evaluated. Fig. 2a and 2b show the time domain emulated MPPG and FPPG signals respectively. The mixed signal as detected from the first MA probe, considered as the first applied input to the ICA model, is depicted in Fig. 2c. The power spectral densities are depicted in Fig. 3. It is clearly shown from Fig. 2 that maternal's fundamental frequency centered at 1 Hz (60 BPM) whereas fetal's one centered approximately at 2.7 Hz (162 BPM). Obviously, in this experiment, the heart rates for both subjects were almost identical which is not the general case. It can also be inferred from Fig. 3c that the $S_F N_M R \approx 0$ dB where the dominant peaks of both the maternal and fetal have roughly the same PSD level. Considering $S_F N_M R$ = 0 dB and there is no any a motion artifact or instrumentation noise, the extracted signals for both maternal and fetal are shown in Fig. 4 and their power spectral densities are shown in Fig. 5. It can be seen from Fig. 4 and 5 that the extracted fetal signal is very clear which proves the applicability of ICA algorithm in this case.

(B) Minimum Applicable $S_F N_M R_{ICA}$

Since the result from the typical case model was excellent, so the next step is to investigate the theoretical $S_F N_M R_{ICA}$ where fetal peaks are still detectable. To accomplish this aim, a series of simulations were conducted to test the performance of the algorithm with different conditions and with a variety in the input $S_F N_M R_{ICA}$ values to ICA algorithm.

Table1 shows the simulation series of this task. Here, the evaluation of an extracted signal quality is done by observation.

It is clearly shown from Table 1 that ICA well-performed without considering the effect of instrumentation noise or any other sort of noises. In this case, the limit of the operation starts with $S_F N_M R_{ICA}$ = -276 dB. Thus, theoretically ICA can extract a clear FPPG with minimum $S_F N_M R_{ICA}$ = -276 dB, as illustrated in

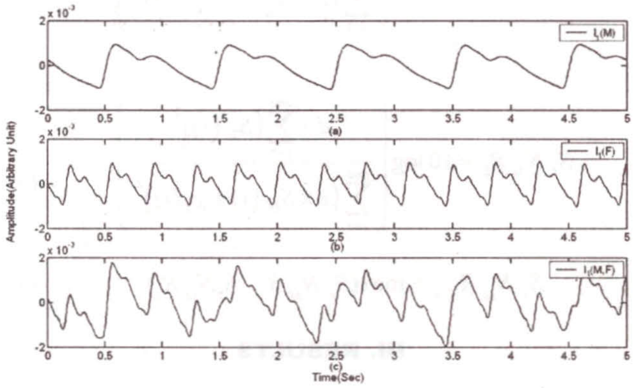

Figure 2: Normalized Time Domain Signals: (a) Emulated MPPG, (b) Emulated FPPG, (c) Mixed Signal (MAPPG)

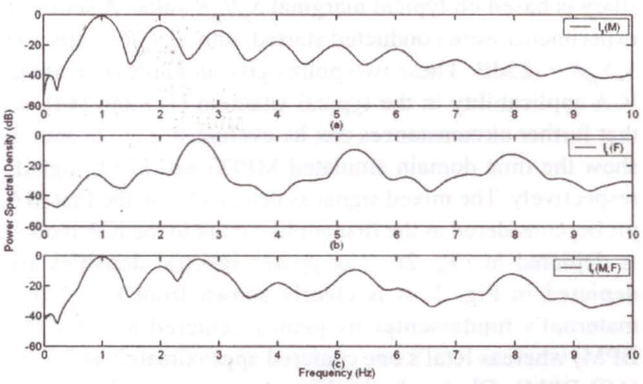

Figure 3: Normalized Power Spectral Densities (Welch Method): (a) Emulated MPPG, (b) Emulated FPPG (c) Mixed Signal (Maternal Abdomen PPG)

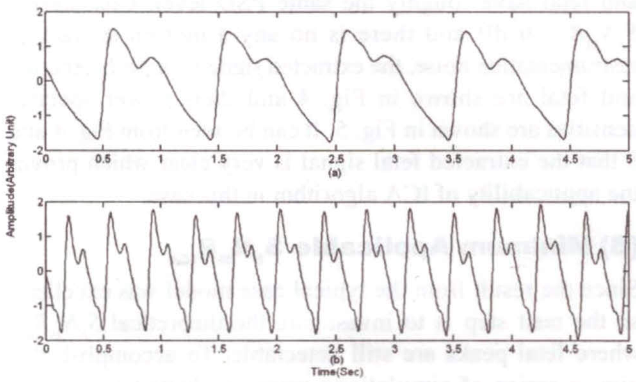

Figure 4: Time Domain Output: (a) $MPPG_{ICA}$, (b) $FPPG_{ICA}$

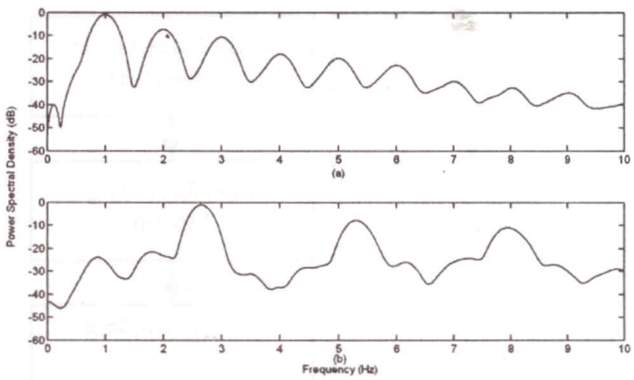

Figure 5: Normalized Power Spectral Density (Welch Method):(a) Estimated $MPPG_{ICA}$, (b) Estimated $MPPG_{ICA}$

Table I
Simulation Series of Minimum $S_F N_M R_{ICA}$ where Fetal Peaks are Still Detectable

Simulation Series	$S_F N_M R_{ICA}$	*Extracted Signal Quality*	*Comments*
ICA_1	0	high	no instrumentation noise
ICA-_2	-25	high	no instrumentation noise
ICA_-3	-13	low	20 dB (AWGN)
ICA_4	-91	very good	no instrumentation noise
ICA_5	-183	good	no instrumentation noise
ICA_6	-276	acceptable	no instrumentation noise
ICA_7	-343	low	no instrumentation noise
ICA_8	-388	no signal	no instrumentation noise

Fig 6. Beyond this limit, ICA does not function well and the detected signal does not give a clear interpretation due to the weak separation as shown in Fig. 7 and 8. Comparing the minimum $S_F N_M R_{ICA}$ in this experiment (-276dB) to the previous method using adaptive noise cancellation (ANC) where the minimum $S_F N_M R_{ANC}$ is -34 dB [14] shows that the proposed method has a superior performance in terms of $S_F N_M R$. However, it is interesting to mentioned that, in our experiment, ICA observed to be very sensitive against instrumentation noise which causes with rapid deterioration of the extracted signals even when $S_F N_M R_{ICA}$ = –13 dB. This reflects the fact that that ICA

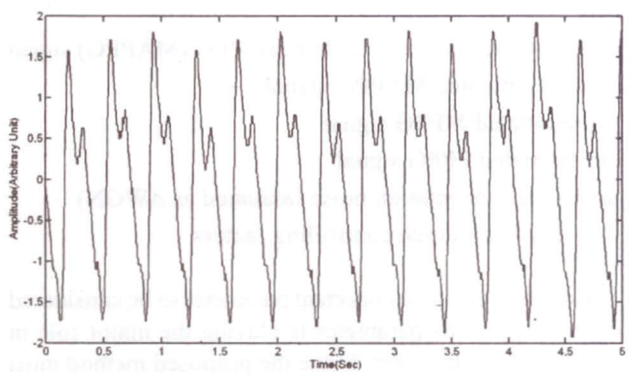

Figure 6: Estimated FPPG ($S_F N_M R_{ICA}$ = -276 dB)

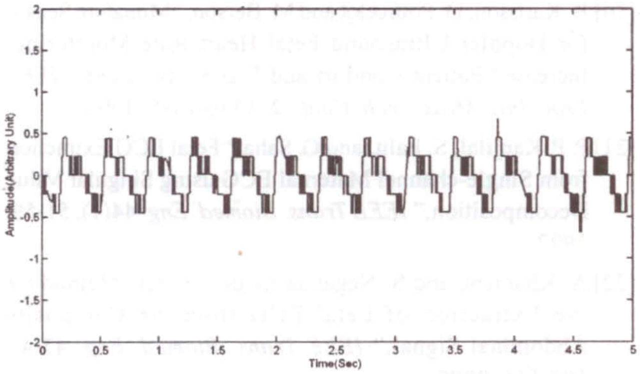

Figure 7: Estimated PPGF ($S_FN_MR_{ICA}$ = -343 dB)

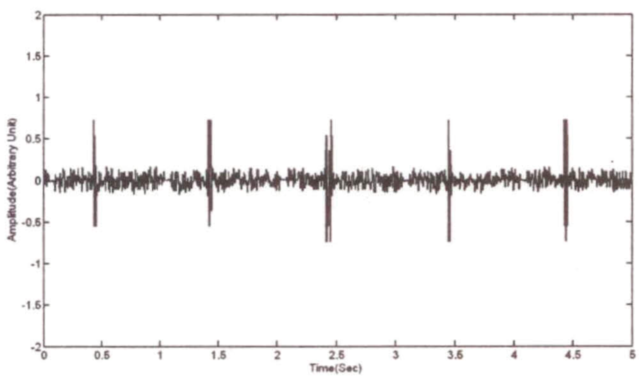

Figure 8: Estimated FPPG ($S_FN_MR_{ICA}$ = -388 dB)

performs a blind extraction of two or more signals without performing any filtering process. Practically, instrumentation noise passes with more than on filtering stage during the preprocessing and will never reach the ICA model in the same level as was assumed in the our experiment. Hence, instrumentation noise must be tackled in the preprocessing steps.

IV. CONCLUSION

In this paper, an investigation of ICA to FHR detection using emulated PPG signals from two independent subjects is presented. Results show that, without considering the effect of any noise source, the minimum applicable $S_FN_MR_{ICA}$= -276 dB. However, it was shown that adding instrumentation noise to the emulated signals deteriorates ICA performance even with very strong fetal peaks. This happens due to the fact that ICA does not perform a filtering process but separates the independent sources blindly. Therefore, a special care must be taken into account in the preprocessing steps in order to assure ICA applicability.

Even though ICA (AMUSE algorithm) was shown an encouraging results in terms of signals extraction, the algorithm has to be reevaluated under further considerations, such as, the effect of motion artifacts and quantization noise. In addition, a comparison between ICA performance using emulated and real PPG signals may also be another issue for investigation.

References

[1] J. G. Webster, *Design of Pulse Oximeters*. IOP Publishing, Bristol and Philadelphia, 1997.

[2] J. Allen, "Photoplethysmography and its Application in Clinical Physiological Measurement," *J. Physiol. Meas.*, **28**, 2007, R1-R39.

[3] J. L. Reuss, and D. Siker, "The Pulse in Reflectance Pulse Oximetry: Modelling and Experimental Studies," *J. Clin. Monit. Comput.*, **18**, 2004, 289-299.

[4] A. Cichocki and S. Amari, *Adaptive Signal and Image Processing*. England: John Wiley & Sons, 2002.

[5] J. TB. Moyle, *Pulse Oximetry*, 2nd ed., BMJ Books, 2002.

[6] M. Nitzan and H. Taitelbaum, "The Measurement of Oxygen Saturation in Arterial and Venous Blood," *IEEE Inst. & Meas. Mag.*, **11**(3), 9-15, 2008.

[7] G. Vishnoi, A. H. Hielscher, N. Ramanujam, *et al.* "Photon Migration through the Fetal Head in Utero using Continuous Wave, Near Infrared Spectroscopy: Development and Evaluation of Experimental and Numerical Models," *J. Biomed. Opt.*, **5**(2), 163-172, 2000.

[8] N. Ramanujam, G. Vishnoi, A. H. Hielscher, *et al.* "Photon Migration through the Fetal Head in Utero using Continuous Wave, Near Infrared Spectroscopy: Clinical and Experimental Model Studies. *J. Biomed. Opt.*, **5**(2), 173-184, 2000.

[9] A. Zourabian *et al.* "Trans-abdominal Monitoring of Fetal Arterial Blood Oxygenation using Pulse Oximetry," *J. Biomed. Opt.* **5**(4), 391-405, 2000.

[10] A. Zourabian and A. B. David "Theoretical Consideration to Optimize Trans-abdominal of Fetal Arterial Blood Oxygenation using Pulse Oximetry," *Pro. SPIE Optical Tomography and Spectroscopy of Tissue IV.*, **4250**, 82-92. 2001.

[11] A. N. Vintzileos *et al.* "Transabdominal Fetal Pulse Oximetry with Near-infrared Spectroscopy," *Am. J. Obst. Gyn.*, **192**(4), 129-133, 2005.

[12] K. B. Gan, E. Zahedi and M. A. Mohd. Ali, "Fetal Heart Rate Detection using Photoplethysmography Technique," in 2006 *Proc. SPS Fkej. UKM*, 175-178, 2006.

[13] K. B. Gan, E. Zahedi, and M. A. Mohd Ali "Feasibility of Fetal Photoplethysmography Signal Extraction using Adaptive Noise Cancelling," in 2007 *Proc. IFMBE Conf.*, **15**, 387-390.

[14] E. Zahedi, G. K. Beng, "Applicability of Adaptive Noise Cancellation to Fetal Heart Rate Detection using Photoplehtysmography" *J. Comp. Biol. Med.*, **38**, 2008, 31-41.

[15] K. B. Gan, E. Zahedi and M. A. Mohd Ali, "Two Channel Abdominal PPG Instrument," in 2008 *Proc. IFBM Conf.*, **21**, pp. 691-693.

[16] A. Nagre, and Y. Mendelson, "Effects of Motion Artifacts on Pulse Oximeter Readings from Different Facial Regions," in 2005 *Proc. Bioeng. Conf.*, pp. 220-222.

[17] W. Johnston, Y. Mendelson, "Extracting Heart Rate Variability from a Wearable Reflectance Pulse Oximeter," in 2005 *Proc. Bioeng. Conf.*, pp. 157-158.

[18] G. A. Dildy, S. L. Clark and C. A. Loucks, "Intrapartum Fetal Pulse Oximetry: Past Present, and Future," *Am. J. Obst. Gyn.*, **175**(1), 1-9, 1996.

[19] F. S. Najafabadi, E. Zahedi and M. A. Mohd Ali, " Fetal Heart Rate Monitoring based on Independent Component Analysis," *J. Comp. Biol. Med.*, **36**, 241-252, 2006.

[20] B. Karlsson, D. Pourcelot and M. Berson, "Miniature Sensor for Doppler Ultrasound Fetal Heart Rate Monitoring. Increased Patient Comfort and Ergonomy in use," *IEEE Proc. Inst. Meas. Tech. Conf.*, **2**, 1336-1338, 1996.

[21] P. P. Kanjilal, S. Palit, and G. Saha, "Fetal ECG Extraction from Single-channel Maternal ECG using Singular Value Decomposition," *IEEE Trans. Biomed. Eng.* **44**(1), 51-59, 1997.

[22] A. Khamene and S. Negahdaripour, "A New Method for the Extraction of Fetal ECG from the Composite Abdominal Signal," *IEEE Trans. Biomed. Eng.* **47**(4), 507-516, 2000.