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Research paper



Linear phase FIR filter to compute fetal heart rate variability

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Abstract

Continuous monitoring of fetal heart rate (FHR) can detect the well-being of the fetus and thus indicates non-reassuring fetal status. Invasive fetal electrocardiography (FECG) using the fetal scalp electrode applied to the fetus scalp allows accurate detection of fetal QRS (FQRS) complexes, however with a risk of infection to the fetus. We have proposed a non-invasive fetal heart rate (NIFHR) filtering technique employing finite impulse response (FIR) filters. We applied Fast Fourier Transform (FFT) to the Physionet abdominal ECG (aECG) records and derived the fiduciary edges of the spectrum of the FECG. A FIR band pass filter (BPF) is designed which is a composite filter consisting of a high pass filter (HPF) followed by a low pass filter (LPF) in that order. The cut off frequencies of these composite filters are the fiduciary edges of the fetal electrocardiography spectrum. A FQRS detector to obtain fetal heart rate variability (FHRV) processes the FQRS signal filtered through these composite FIR filters. It is observed that channel 4 from records r01 and r08 obtained 100% results for sensitivity, positive predictive value and accuracy while, the overall accuracy was 92.21%. This design can also be extended to compute maternal heart rate.

Keywords: *Abdominal ECG; Composite Linear Phase FIR Filter; Fetal ECG; Fetal Heart Rate; QRS Detector*.

1. Introduction

Five years later after Einthoven first discovered an electrical activity in a human heart, Cremer identified fetal electrocardiograph (FECG) from the abdominal and vaginal set of electrodes. Winkler discovered six years hence [1] that a fetal heart rate (FHR) higher than 160 bpm and less than 120 beats per minute (bpm) indicated fetal distress. Decelerations of the FHR predicts fetal hypoxia. WHO Media Centre 2012 reported congenital birth defects of approximately [1] in every 33 infants and 3.2 million birth defects. In the same year, it was reported that the Infant Mortality Rate (IMR) in India was about 11 deaths per 1000 live births [2]. The trend of birth defects leading to death in infants is seen till today. Therefore, the monitoring of FHR is sensitive and one can detect fetal asphyxia before the phase of fetal cell damage. Elec- tronic FHR monitoring was therefore a very good attempt to elim- inate the effects of fetal asphyxia. Presently, the prominent meth- ods to monitor fetal well-being are cardiotocography (CTG) and Ultra sound transducers.

Fetal distress also known as non-reassuring fetal status is a complication during pregnancy or labor. It can take place due to placenta abruption, umbilical cord prolapse, breathing problems etc [3]. The various non-reassuring patterns seen on a CTG to indicate such fetal distress are increased or decreased fetal heart rates especially during and after a contraction, decreased variabil- ity in FHR and late decelerations. FHR changes obtained using electrocardiogram (ECG) signal are more accurate compared to the recordings taken from the CTG [4].

Non Invasive FECG (NIFECG) signals obtained from the ab- dominally placed ECG electrodes show the main sources of noise to be the maternal ECG (MECG), low frequency baseline wander, 50/60 Hz power line interference (PLI), maternal electromyogram and other artifacts [5]. The MECG amplitude is more than four times the NIFECG and is a challenge to extract and separate the NIFECG from the MECG [6]. Invasive FECG using the scalp electrode applied to the fetal scalp allows easy detection of fetal QRS complexes. These signals have much larger amplitudes than abdominally obtained signals. There is however a risk of infection to the fetus [7]. Detection of NIFECG signal and associated analy- sis is a very powerful and advanced method in clinical diagnosis and biomedical applications. The NIFECG contains potentially valuable information that assists clinicians to make appropriate and timely decisions, especially during labor and the last weeks of the 3rd trimester. There is a limitation to perfectly monitor fetal heart rates, improving the signal to noise ratio (SNR) of the NI- FECG. Some methods for extracting NIFECG need prior infor- mation for extraction, others need signals from many leads and some require recording from the mother's thoracic area. These methods require electrodes to be placed all over the mother's body and thoracic area, making it inconvenient in non-clinical environ- ments such as the maternal home care monitors [8]. A technique should therefore be developed to allow extraction of FECG with a minimal abdominal lead setup that is independent of any recording technique. A single-lead configuration is helpful for implementa- tion in a mini health care unit, making it suitable for ambulatory and long-term monitoring. [9].

1.1. Maternal and fetal frequency bands

The maternal adult heart rate normally ranges from 70-to110 bpm based on the condition of the patient, while the FHR varies with the gestational period. The FHR of the fetus can be recorded right from the 20th week of pregnancy [11]. It is very well known that the insulating layer known as vernix caseosa becomes the block- ing factor to record good quality FECG signals during the 28th week to 32nd week. The FECG signal becomes lesser than 10 µv during these weeks [12] as shown in Figure 1. Therefore it is ad- vised not to record FECG signals during the formation of this layer, however

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the layer is dissolved in the 37th to 38th week [13]. To obtain an individual QRS complex of the MECG and FECG is quite a challenging task especially when (i) the maternal QRS amplitude (MQRSA) is about 4 to 20 times larger than the fetal QRS amplitude (FQRSA) [14] and (ii) the MECG and FECG over- lap both in time and frequency domain [7].

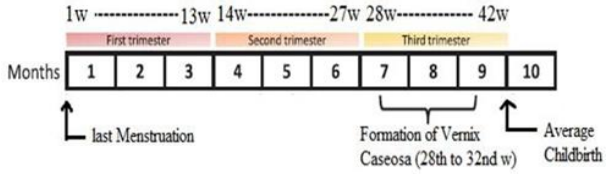


Fig. 1: Prenatal Development with Respect to Fetal Monitoring (W = Week).

Table 1: Maternal And Fetal Parameters: QRS Amplitude, QRS Width and QRS Frequency Bandwidth

Time Domain			
MQRSA	FQRSA	MQRSW	FQRSW
100 - 150 µv [7]	60 µv [7]	120 ms [15]	80 ms [15]
150 µv [6]	30 µv [6]	100 ms [6]	50 ms [6], 53 ms [16]
300 µv [17]	10-20 µv [17]	100 µv [14]	3- 25µv [14]

Separating FECG from MECG gets tougher when the abdominal ECG (aECG) is engulfed with noise such baseline wandering, PLI and other high frequency noise. Abdominal ECG signals can be obtained by normally placing a set of surface electrodes over the mother's abdomen in the third trimester (after the 36th gestation week) and having prior knowledge of the fetus presentation by using the ultrasound procedure. A review of the various abdominal electrode placements can be obtained in [2, 9, 18, 29]. To obtain the MQRS and FQRS frequency spectrum, a literature survey was compiled. The recorded values for the following parameters are: maternal QRS amplitude (MQRSA), fetal QRS amplitude (FQR- SA), maternal QRS width (MQRSW), fetal QRS width (FQRSW), MQRS frequency bandwidth (MQRSBW) and FQRS frequency bandwidth (FQRSBW) as shown in Figure 2. However just knowing the QRS width does not tell us the frequency spectrum of that ECG signal.

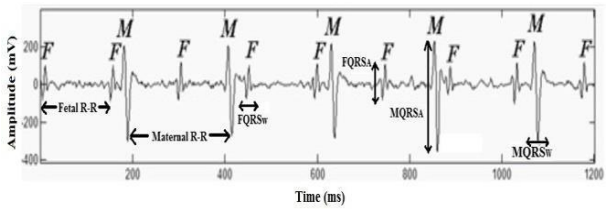


Fig. 2: Interleaved Maternal and Fetal ECG Signals.

Table 2: Maternal and Fetal Parameters: QRS Frequency Bandwidth

Frequency domain	
$MQRS_{BW}$	$FQRS_{BW}$
18 – 35 Hz [15]	27-53 Hz [15]
10 - 40 Hz [17], 10 - 25Hz [20], 10 – 30 Hz [14]	Starts at 20 Hz [17]
5 – 15 Hz [10] , 2 - 20 Hz [21]	15 – 40 Hz [22]
1-30 Hz [23]	20 – 60 Hz [14]

Frequency domain
MQRSBW QRSBW
18 – 35 Hz [15] 27-53 Hz [15]
10 - 40 Hz [17], 10 - 25Hz
[20], 10 – 30 Hz [14] Starts at 20 Hz [17]
5 – 15 Hz [10], 2 - 20 Hz
[21] 15 – 40 Hz [22]
1-30 Hz [23] 20 – 60 Hz [14]

As seen in Table 1 and Table 2, the maternal beats per minute normally ranges from 50 to 210 bpm with an average of 80 or 89 bpm [14, 15]. While the FECG bandwidth ranges from 0.05 – 100 Hz [19], the fetal beats per minute is recorded by various authors such as 60-240 bpm [14], 120 – 160 bpm [15], 110 – 160 bpm
[16] and 78 – 210 bpm, where mostly all have taken an average value as 140 bpm.

In our experiment to extract MQRS and FQRS from databases to find the average frequency spectrum , we used the two Physio- net databases [24] firstly, the non-invasive fetal electrocardiogram database (nifecgdb), with a sampling frequency of 1KHz , fifty- five abdominal recordings (five channels each) taken from a single mother having gestation age of 21-41 weeks. Channels 1 and 2 refer to the maternal thoracic ECG signal as MQRS reference signal. The second abdominal database is the abdominal and direct fetal electrocardiogram database (adfecgdb) with a sampling fre- quency of 1 KHz, four channel abdominal recordings of 5 minutes each from five mothers in labor having gestation age of 38-41 weeks. The direct fetal scalp ECG electrode (channel one) gives the FQRS as reference. To compare the MQRSBW and FQRSBW band pass frequencies for the maternal thoracic and fetal scalp ECG signals, we marked the Q-R-S fiducial edges for MECG and FECG of the above databases. The Fast Fourier transform (FFT) for the maternal and fetal QRS band pass frequency spectrum were averaged at MQRSBW to be (10 – 34 Hz) and FQRSBW ranged from (20 – 56 Hz).

2. Methodology composite linear phase FIR filter design

In this paper, we proposed a method to obtain non-invasive fetal heart rate (NIFHR) from a single lead maternal abdominal signal by applying the composite high and low pass digital FIR filters in tandem. An FQRS detector is also used based on Pan Tomkins QRS detector [10] using an amplitude squaring module, appropri- ate moving window integrator and an adaptive threshold to effec- tively obtain the fetal heart rate variability (FHRV). From the literature reported in Table 1 and Table 2, we can design a high pass FIR filter with a cut-off of 35 Hz which is the lower fiduciary edge of the fetal spectrum that will filter low frequency noise in- cluding baseline wander frequencies and also avoid maternal spec- trum overlap. This is followed by a low pass FIR filter with a cut- off of 48Hz which is the upper fiduciary edge of the spectrum of FECG. This cut off frequency will remove the PLI frequency at 50Hz and the PLI harmonics and other high frequency noise. To effectively extract the required fetal information from the aECG following composite FIR filters were designed.

2.1. Model and design: linear phase FIR high pass filter

In this section, the design of the FIR high pass filter with linear phase is modelled for three regions of the filter response H (ω) as shown in Figure 3.

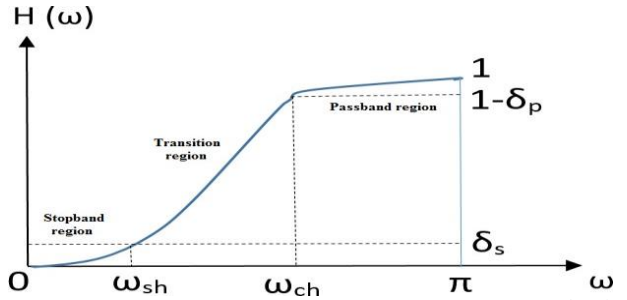


Fig. 3: Linear Phase FIR HPF Model Magnitude Response H (Ω) Showing the Three Regions.

Region 1 (stopband):
$H(\omega) = \delta_s[1 - \cos(k_{sh}\,\omega)]$ $0 \le \omega \le \omega_{sh}$

Region 2 (transition) :
$H(\omega) = \delta_s + (1-\delta_p-\delta_s)\sin\left|k_{th}(\omega-\omega_{sh})\right|$ $\omega_{sh} \le \omega \le \omega_{ch}$

Region 3 (passband):
$H(\omega) = (1-\delta_p) \pm \delta_p \sin\left[k_{ph}(\omega-\omega_{ch})\right]$ $\omega_{ch} \le \omega \le \pi$ (1)

Using equation (1), the filter design parameters ksh, kth and kph for the 3 regions of the HPF can be evaluated, where; ωsh is the stopband edge frequency while ωch is the cut off frequency in the passband. δs and δp are the stopband attenuation and passband ripple respectively.
The impulse response coefficients h(n) for the HPF FIR can be obtained by computing the integral limits of the three regions as shown in the filter model magnitude response in Figure 3.

2.2. Model and design: linear phase FIR low pass filter

Similarly, in this section, the design of the FIR LPF with linear phase is modelled for three regions of the filter response H (ω) as shown in Figure 4. As in HPF design, ω is the frequency variable, H (ω) is the magnitude of the filter response, δs is the stopband attenuation and δp is the passband ripple. kpl, ktl and ksl are the passband, transition and stopband filter design parameters respectively.

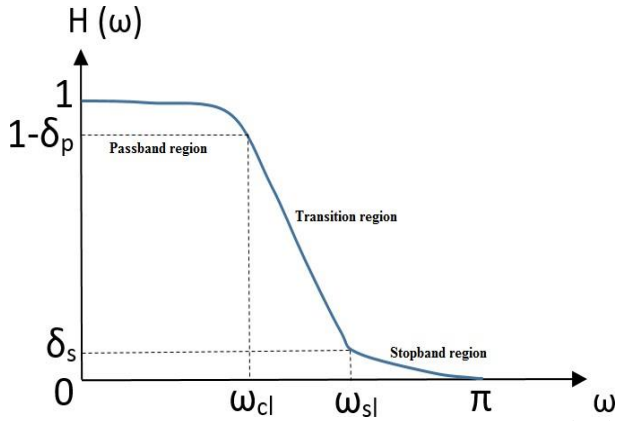


Fig. 4: Linear Phase FIR LPF Model Magnitude Response H (Ω) Showing the Three Regions.

Region 1 (passband):
$H(\omega) = (1-\delta_p) + \delta_p \cos(k_{pl}\,\omega)$ $0 \le \omega \le \omega_{cl}$

Region 2 (transition):
$H(\omega) = \delta_s + (1-\delta_p-\delta_s)\cos\left|k_{tl}(\omega-\omega_{cl})\right|$ $\omega_{cl} \le \omega \le \omega_{sl}$

Region 3 (stopband):
$H(\omega) = \pm\delta_s \pm \delta_s \sin\left[k_{sl}(\omega-\omega_{sly})\right]$ $\omega_{sl} \le \omega \le \pi$ (2)

Using equation (2), the filter design parameters kpl, ktl and ksl for the three regions of the LPF can be evaluated, where, ωsl is the stopband edge frequency while ωcl is the cut off frequency in the passband.
With reference to the four cases of linear phase FIR filters as listed

3. Results

The designed linear phase FIR filters used in tandem were high pass filters followed by the low pass filter to form a bandpass filter as shown in Figure 5 as compared to other similar linear phase FIR band pass filter designs such as [26,27]. The following cut off frequencies for the FIR filters were substituted.

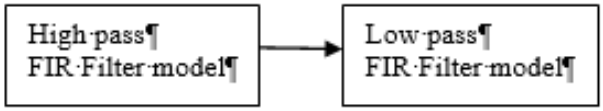


Fig. 5: Linear Phase FIR Band Pass Filter Model.

For FIR HPF: ωsh = 35 Hz and ωch = 36 Hz, while FIR LPF FIR : ωcl = 48Hz and ωsl = 49Hz. The measurement of the magnitude response of these filters are compared in Table 3 and 4 with the filter design specifications. Table 5 and Table 6 depict the performance of the filter for various filter order (N). There is reduction of Gibb's phenomenon using our filter designs.

Table 3: High Pass Fir Filter Specifications Along with Measured Magni- Tude Response Values

Linear phase FIR HPF	Pass band edge (ωch) rad/s	Stop- band edge (ωsh) rad/s	Transition bandwidth (ωch - ωsh) rad/s	Max. pass- band loss (dB)	Min. stopband attenuation (dB)
Design specs.	56 π	54π	2π	±0.873	40
Measured specs.	57.3π	52π	5.3π	+0.28, -0.183	38

Table 4: Low Pass FIR Filter Specifications Along with Measured Magni- Tude Response Values

Linear phase FIR LPF	Pass band edge (ωcl) rad/s	Stop band edge(ωsl) rad/s	Transition bandwidth (ωsl - ωcl) rad/s	Max. pass band loss (dB)	Min. Stopband attenuation (dB)
Design Specs.	96 π	98 π	2 π	± 0.873	40
Measured Specs.	95.5π	99.86 π	4.36π	-0.132	40

Table 5: Variations of Passband Loss and Stopband Attenuation for HPF with Various Filter Orders

Filter order (N)	200	500	1000	1500	2000
Passband loss (dB)	0.534	0.164	0.284	0.044	0.076
Stopband attenuation (dB)	21.04	30.56	40	40.6	42

Table 6: Variations of Passband Loss and Stopband Attenuation for LPF with Various Filter Orders

Filter order (N)	200	500	1000	1500	2000
Passband loss (dB)	0.569	0.442	0.132	0.1277	0.1264
Stopband attenuation (dB)	25.59	32.46	40	40.7	41.3

4. Performance analysis of the fetal QRS detector

The fetal R-peaks generated by our linear phase FIR filter algorithm was compared with the scalp fetal peak annotations from the website. We used the distance based measure to assess the error for the fetal detection between the true R-peak location and the detected fetal R-peak using our algorithm. As per ANSI/AAMI guidelines (ANSI/AAMI/ISO EC57 1998/(R) 2008) [9, 28], sensi- tivity (Se), positive predictive value (PPV) and F1 (accuracy) are the classical statistics for evaluating QRS detectors as shown in equations (3) to (5) where, TP, FN and FP are True Positive (cor- rectly identified fetal peaks), false negative (missed fetal peaks) and false positive (falsely identified peaks), respectively. For ex- ample, we evaluated our algorithm for the adfecgdb database for the one minute record (r01) of channel 4 and found the TP = 129; FN = 0 and FP = 0. The sensitivity, PPV and the accuracy was obtained to be 100%. The average FHR values for the true refer- ence and algorithm FHR were computed to be 128.98 bpm [29]
and 129 bpm, respectively. The Figure 6 illustrates the true reference FHR bpm plotted with our algorithm based FHR. The dotted lines indicate the ± 5 bpm with respect to the true reference FHR trace. It was seen that the difference between the reference FHR and algorithm FHR was less than the ± 4 bpm. All simulations were implemented using the Matlab tool. The five records from the Physionet adfecgdb database [30] were evaluated using our proposed filter design and the modified Pan Tomkins FQRS detec- tor algorithm as seen in Figure 7. It is observed that channel 4 from records r01 and r08 obtained 100% results for Se, PPV and F1. It was seen that the algorithm missed a large fetal R-peaks for records r04 and r07.

$$Se = \frac{TP}{TP + FN} \tag{3}$$

$$PPV = \frac{TP}{TP + FP} \tag{4}$$

And

$$F_1 = 2.\left(\frac{Se.PPV}{Se + PPV}\right) \tag{5}$$

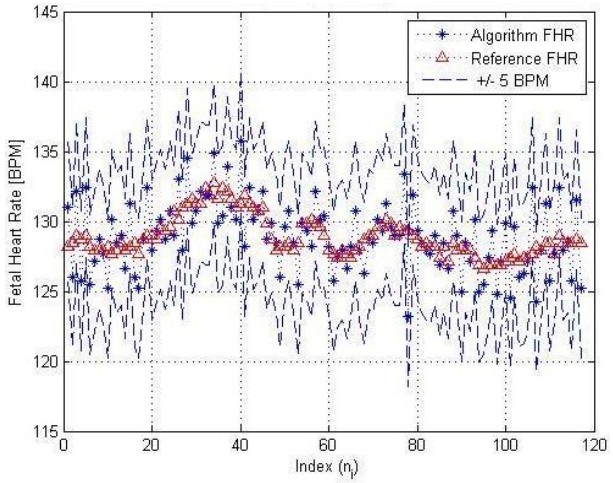


Fig. 6: Illustration of the True Reference FHR (Direct Scalp ECG Plotted with Our Linear Phase FIR BPF Algorithm Computed FHR for Record R01 of Adfecgdb (Channel Four) for One Minute Trace. Blue Dotted Lines Indicate The ±5 BPM Limits with respect to the Reference FHR Trace.

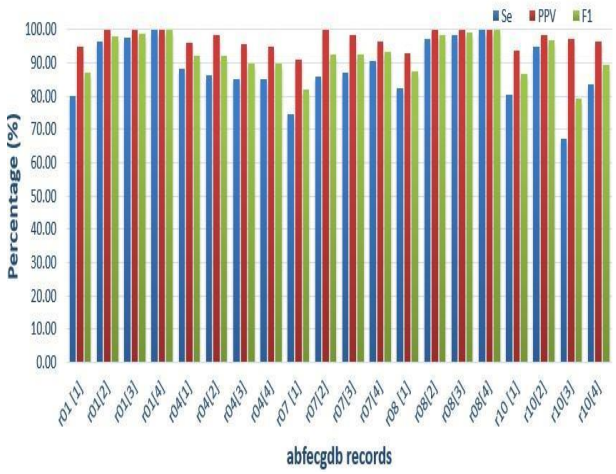


Fig. 7: Graphical Representation of the Evaluation of the FQRS Detection Using the Composite Linear Phase FIR BPF for the Adfecgdb Records.

5. Conclusion

In this paper, we proposed a novel non-invasive fetal heart rate filtering technique, which employed a linear phase FIR BPF containing a tandem of a high pass, and low pass FIR filters. The cut off frequencies of these composite filters are estimated from the Physionet aECG signals. We designed both the FIR filters such that the transition region width of each is 1 Hz, which improved the accuracy of the FHR determination. The fetal R peaks generated by our algorithm were compared with the scalp fetal peak annotations from the Physionet database. The fetal R peaks were found to be in close agreement (mostly for channel 4) with each other including the average FHR values between the true reference and algorithm FHR. Similarly, other performance indices such as sensitivity, PPV and F1 for the most records of the adfecgdb database were found to be having promising results. The overall accuracy was computed to be 92.21%. Our linear phase composite FIR filters can be extended to find the maternal heart rate. As future scope, we can apply our proposed FIR filters to real time aECG signals to extract FHR thereby, monitoring fetal health status [31]. The fetal health condition can be wirelessly transferred to the mother as well as the clinicians using mini smart phone monitors.

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