Effect of respiration harmonics on beat-to-beat analysis of heart signal

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***Abstract* — Doppler radar is a promising tool for non-invasive physiological monitoring. However, its output signal is affected by respiration harmonics interference due to inherent respiratory features and a radar detection method. Respiration harmonics present a challenge for sophisticated beat-to-beat analysis of heart signals for extraction of heart rate variability (HRV) parameters. This paper discusses causes of these harmonics along with proposing the utilization of maximal overlap discrete wavelet transform (MODWT) for minimizing their interference. The results are presented in terms of average heart rate and the standard deviation of interbeat intervals (SDNN). Adaptive notch filter is used for comparing the effect of harmonics cancellation on beat-to-beat analysis of heart signals. Compared to the results of the notch filter, the SDNN difference from MODWT is lowered by 5.12% to 91.43%.**

***Keywords* — Respiration harmonics, Quadrature Doppler radar, Linear demodulation, Maximal overlap discrete wavelet transform, IIR notch filter, beat-to-beat analysis.**

I. INTRODUCTION

With the fast advent of physiological sensing techniques, Doppler radar (DR) has become a very popular choice for remote sensing of important vital signs such as heart rate, heart rate variability (HRV), respiration rate, tidal volume, etc. This form of sensing method utilizes the phase modulation effect caused by human physiological movement. However, the use of DR signals is greatly affected by harmonics interference caused by respiration signals. The amplitude of the respiratory signals ranges from 4 to 12 mm with a normal respiratory rate between 5-25 breaths per minute. Whereas, heartbeat amplitude ranges from 0.2 to 0.5 mm with a normal heart rate between 50-120 beats per minute. Since the amplitude of the respiratory signals is much larger than the heartbeat signals, respiratory harmonics can cause interference with the identification of heartbeat signals. Especially, when the respiration harmonics and heartbeat signal are very closely located, the harmonics might be interpreted as the heartbeat signal mistakenly [1]. Moreover, if the harmonics are of comparable amplitude to the heartbeat signals, it will affect the correct detection of peaks making the radar signal unsuitable for sophisticated analysis like HRV. To overcome this obstacle, various studies have proposed different methods for respiration harmonics removal. For this purpose, Huang et. al. proposed using an adaptive harmonic comb notch filter which consists of multiple notch frequencies at respiration and its harmonics with a magnitude determined by the ratio of harmonic amplitude [2]. A study presented respiration harmonic cancellation by employing complex signal demodulation without any DC offset calibration [3]. To suppress the respiration harmonics, Hough transform accumulating signal energy along a preset frequency trajectory for extracting heartbeat signals has also been used [4]. Although there have been studies on the improvement of average heart rate through respiration harmonics cancellation, its effect on the beat-to-beat analysis of heart signals is still unexplored. In this paper, the causes of respiration harmonics are discussed in Section II. Furthermore, the compounded effect of respiration harmonics on beat-to-beat analysis of heart signals is presented. Maximal Overlap Discrete Wavelet (MODWT) analysis is utilized for the cancellation of such harmonics and facilitating the beat-to-beat analysis of heart signals. It was shown that the notch filters used previously for harmonics cancellation can improve the average heart rate calculation but are not enough for beat-to-beat analysis of heart signals. Section IV. discusses the improvement in such analysis achieved with the proposed method. Among the contact sensors, photoplethysmography (PPG) signal is becoming widely popular due to its low cost, ease of usage, and reliability. Hence, to compare the results in terms of average heart rate and standard deviation of interbeat intervals from the proposed method, the PPG signal is used as a reference.

II. RESPIRATION HARMONICS

The causes behind respiration harmonics in Doppler radar measurements can be broadly classified into two reasons: due to inherent features of respiration, and due to the non-linearity of the radar output signal. First, this type of harmonics can be caused by the respiratory movement itself if it is not an ideal single-tone periodical movement [3], [5]. The posture of a subject during data collection with radar is another factor that contributes to respiration harmonics. When a person breathes, there is micro-motion in both the abdomen and thorax. Through a respiratory motion analysis, Quirk et. al. reported abdominal displacement to be four times the thoracic displacement [6]. For certain postures such as sitting, this abdominal movement may dominate over that of the chest [7]. As a result, stronger or more respiration harmonics contents may be found in such data. [3]. Figure 1. shows the frequency domain plots of respiratory signals obtained after filtering the signals from 0.1 Hz to 1 Hz. Figure 1 (a) illustrates the frequency domain plot of a single channel radar signal where the respiration harmonics are present partly as a radar artifact. Additionally, Figure 1(b)-(d) showing PPG and accelerometer

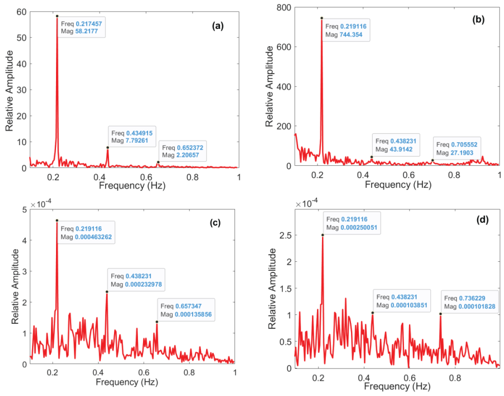

Fig. 1. Frequency domain plots of (a) Radar, (b) PPG, (c) X-axis accelerometer, and (d) Z-axis accelerometer respiratory signals.

signals in frequency domain validates these harmonics are a respiratory feature that is present in every signal although in different magnitudes.

The setup of a direct-conversion quadrature Doppler radar is shown in Figure 2. Its baseband output signals can be expressed as-

$$B_I(t) = A_B cos(\theta + \frac{\pi}{4} + \frac{4\pi x(t)}{\lambda} + \Delta\phi(t)) \tag{1}$$

$$B_Q(t) = A_B sin(\theta + \frac{\pi}{4} + \frac{4\pi x(t)}{\lambda} + \Delta\phi(t)) \tag{2}$$

Here, $x(t)$ represents the target's time-varying displacement, the baseband amplitude due to receiver and mixer gain is given by A_B, the constant phase shift, and the residual phase shift is given by θ and $\Delta\phi(t))$ respectively. Due to the non-linearity of cosine and sine functions, even if the respiratory signal were a perfect sinusoid, Doppler radar baseband outputs would still contain harmonics [8]. If linear demodulation is used to combine quadrature outputs, this harmonic content would remain in the output signal. Linear demodulation (LD) is a technique that employs the selection of the major component among in-phase and quadrature channel signals and it can be mathematically expressed as-

$$\hat{x}[n] = d^T r[n] \tag{3}$$

In Equation (3), $x[n]$ is the output of the LD, d^T is the transpose matrix of the eigenvector from the covariance matrix, and $r[n]$ is the matrix composed of the I and Q received signals. Here, the arc is rotated about its center and the channel with the highest peak-to-peak amplitude at a certain degree of rotation is selected [9]. As the channel which is in optimum position in that degree of rotation is selected, the effect of non-linearity still exists in the demodulated signal. Hence, the respiration harmonics are not removed with linear demodulation.

A. *Maximal overlap discrete wavelet transform (MODWT)*

MODWT is a linear filtering operation that transforms a signal into detail coefficients and scaling coefficients with

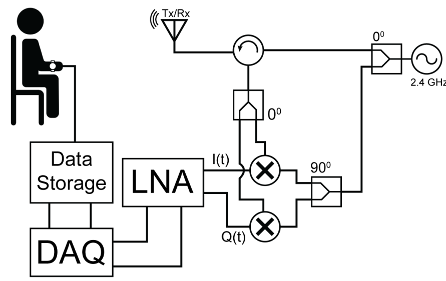

Fig. 2. Block diagram of a quadrature Doppler radar cardiopulmonary measurement system along with the experimental setup.

respect to variations over a set of scales. Although it is similar to discrete wavelet transform, MODWT offers better alignment of decomposed wavelet and scaling coefficients at each level with the original time series. Hence, the analysis of localized signal variation with respect to scale and time is easier with such a process [10]. If $C^{(M)}_{(o,n)} = x_n$ is a time series, then the convolution of MODWT can be written as-

$$d^{(M)}_j, n = \sum_{l=0}^{L-1} \tilde{h}_L C^{(M)}_{j-1}, (n - 2^{j-1}l) modN \tag{4}$$

$$C^{(M)}_{j,n} = \sum_{l=0}^{L-1} \tilde{g}_L C^{(M)}_{j-1}, (n - 2^{j-1}l) modN \tag{5}$$

where, n = 0,1,..., N-1 denotes the length of time-series, $d^{(M)}_{j,n}$ and $C^{(M)}_{j,n}$ are the wavelet and scaling coefficients respectively and M stands for MODWT. The wavelet and the scaling filter are given by $\tilde{h}_L$ and $\tilde{g}_L$ respectively [11].

III. Experimental Setup

Data were collected from 3 healthy male subjects (A-C) under protocol no. 14884 approved by the Committee on Human Studies (CHS) of the University of Hawaii system. During data collection, all the subjects were asked to sit still and breathe at a metronomic rate of 13, 15, and 18 breaths per minute (b/m). The radar was placed 1 meter away from the subject and the single patch antenna was elevated according to the height of the subject's sternum. An E4433B signal generator transmitted a signal with a frequency of 2.4 GHz and an amplitude of 16dBm. The setup included a 90-degree ZX10Q-2-25-S+ splitter, ZFM4212 mixers from Mini-circuits, and a Narda 4923. Next, the I and Q mixers' outputs are low pass filtered at 10 Hz and amplified 20 times by passing the outputs through a DC Coupled SR560 LNA. Finally, the DAQ records the signals with 18-bit resolution and at a 100 Hz sampling rate. MAXREFDES103: Wrist-Based SpO2, HR, and HRV Health Sensor Platform is used to collect the PPG signal from the right-hand wrist of the subject. The watch utilizes a green LED light in transmission mode to measure the volumetric variations of blood circulation [12]. Maxim Integrated software was used for PPG data recording at the

Table 1. Average heart rate (bpm) extraction using the proposed method.

Dataset	With notch filters			With MODWT			Reference
	I-Channel	Q-Channel	LD	I-Channel	Q-Channel	LD	
Subject A (13b/m)	67.8	71.09	68.63	66.57	66.53	66.98	70.34
Subject A (15b/m)	66.38	69.79	67.23	69.74	70.32	70.18	69.16
Subject A (18b/m)	67.94	76.32	67.94	67.45	67.88	67.94	67.95
Subject B (13b/m)	72.27	68.18	81.81	73.22	74.22	74.25	70.34
Subject B (15b/m)	72.27	91.36	90.68	72.99	73.19	73.03	72.60
Subject B (18b/m)	69.57	71.3	71.3	69.72	68.77	68.73	68.75
Subject C (13b/m)	64.91	71.64	71.63	73.01	73.03	72.16	70.09
Subject C (15b/m)	72.79	73.22	72.79	69.74	69.75	69.73	70.60
Subject C (18b/m)	68.77	68.77	68.77	68.83	68.76	67.55	66.67

same sampling rate. To align the data from two different DAQs, both radar and PPG data were recorded with timestamps.

IV. Results

In this section, the effect of respiration harmonics is first presented. All the signals were bandpass filtered from 0.85 to 5 Hz with a filter order of 600. Next, MODWT is applied to reduce the effect of harmonics near the heart fundamental frequency. The MODWT uses Symlet (sym) 4 mother wavelet and decomposes the signal into eight levels of frequency scales. For this case, the level 6 signal is reconstructed as the heart signal. The reconstructed heart signal of the same dataset is presented in Figure 3. Using MODWT on a filtered signal helps to suppress the spectral leakage associated with a bandpass filter. The harmonics of heart frequency are not of considerable amplitude and are comparable to the noise in the filtered signal. Hence, only the fundamental frequency was chosen for our analysis. Since MODWT decomposes the signal into different frequency ranges with energy levels, it helps to determine which decomposition level contains the desired heart fundamental frequency. It also reduces the amplitude of the higher-order harmonics present in the range of the fundamental heart frequency. Usually, third or fourth harmonics are present very near the fundamental heart rate depending on the metronomic breathing rate of the subject. In that cases, it is not possible to remove those harmonics with MODWT. Hence, IIR notch filter can be applied at the respiration harmonic frequencies present after the wavelet transform. Figure 4. depicts the harmonics cancellation using such a filter. The magnitude of the notch is adapted according to the harmonic's amplitude obtained from the FFT plot. Table I. lists the results in terms of the average heart rate obtained with the proposed method. In some cases, MODWT shows closer average heart rates (HR) to the reference than the notch filter such as LD and Q-Channel signals of Subject B at 15b/m. The mean difference of the average HR of I-Channel and reference PPG signal decreases from 1.99 bpm to 1.67 bpm when MODWT is used instead of the notch filter. With the notch filter, the mean average HR difference is 4.39 bpm and 4.62 bpm for Q-channel and LD signals respectively. This mean difference significantly decreases to 1.71 bpm for the Q-Channel signal and 1.40 bpm for the LD signal when MODWT is employed. In previous studies regarding harmonic cancellation, adaptive comb notch filters are popularly utilized [2], [6]. Satisfactory accuracies were achieved in those studies since all of them were only concerned with average heart rate. However, such a filter is not capable of improving the beat-to-beat analysis of heart signals as evident in Table II. The results are presented in terms of the difference between the standard deviation of the beat-to-beat interval (SDNN) of the radar from the mean interval and that of the corresponding PPG signal in milliseconds (ms). SDNN is a widely used technique in beat-to-beat analysis. [13]. Hence, it was chosen to show how beat-to-beat analysis can be facilitated through harmonics cancellation using MODWT instead. From Table II, it can be seen that using MODWT can decrease the difference between the SDNNs of the radar I-channel and reference heart signals by at most 91.43% compared to the notch filtering method. For the Q-channel signal the highest improvement was seen for Subject-B at 13b/m. But, for the same subject at 15b/m and 18b/m, the difference increases between the Q-Channel and reference PPG signal. However, for linear demodulated signal the improvement using MODWT rather than notch filter is consistent (67.58% at most) for all datasets.

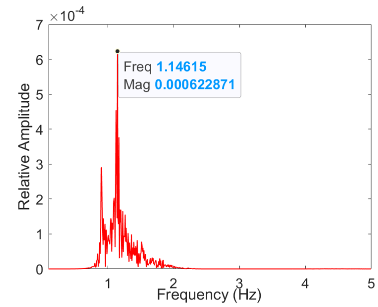

Fig. 3. Level 6 reconstructed linear demodulated signal after applying MODWT for Subject C at 18b/m.

V. Conclusions

This study focuses on the causes of respiration harmonics and its significant impact on the beat-to-beat analysis of heart signals. Based on the observation, the origin of harmonics can be considered mostly as a respiration attribute rather than a demodulation artifact. Single-channel radar can also

Table 2. Difference of SDNN of heart signals from the radar and PPG signal in milliseconds (ms). After harmonics reduction using notch filters versus the proposed method, the latter shows a lower SDNN difference.

Dataset	With notch filters			With MODWT		
	I-channel	Q-channel	LD	I-channel	Q-channel	LD
Subject-A (13b/m)	75.22	182.69	170.81	71.37	145.39	149.54
Subject-A (15b/m)	92.24	141.41	125.6	75.03	109.62	115.01
Subject-A (18b/m)	78.74	181.57	131.99	64.52	141.26	118.46
Subject-B (13b/m)	64.64	65.68	48.9	5.54	21.89	18.56
Subject-B (15b/m)	112.73	71.16	93.9	65.82	82.75	78.92
Subject-B (18b/m)	130.4	93.05	123.12	110.89	100.67	80.71
Subject-C (13b/m)	122.34	99.37	112.03	43.32	38.44	36.33
Subject-C (15b/m)	37.44	74.81	68.77	34.81	36.07	30.32
Subject-C (18b/m)	95.29	95.9	95.15	42.02	44.09	34.88

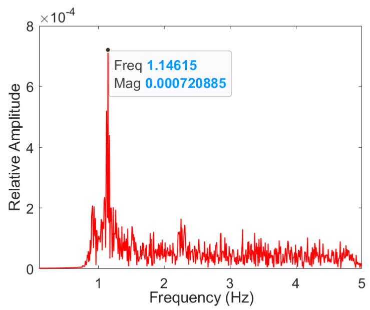

Fig. 4. Respiration harmonics cancellation using adaptive IIR notch filter of linear demodulated heart signal of Subject C at 18b/m.

cause a strong second harmonic in the null point (which may contribute to the fourth harmonic). Previous studies on the such harmonics are limited to exploring its effect on average heart rate only. However, here it was shown that using MODWT can reduce respiration harmonics and significantly improve beat-to-beat analysis of heart signals. In this study it was observed that the accuracy varied from subject to subject which might be due to the inherent body movements. For linear demodulated data the improvement using MODWT was consistent. However, for the single channel signals, some cases showed an increase in SDNN difference from MODWT by as much as 16.29 %. This issue will be explored in future.

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