2022 44th Annual International Conference of
the IEEE Engineering in Medicine & Biology Society (EMBC)
Scottish Event Campus, Glasgow, UK, July 11-15, 2022

Heart rate detection using single-channel Doppler radar system

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2022 44th Annual International Conference of the IEEE Engineering in Medicine & Biology Society (EMBC) | 978-1-7281-2782-8/22/$31.00 ©2022 IEEE | DOI: 10.1109/EMBC48229.2022.9871199

***Abstract*— A number of algorithms have been developed to extract heart rate from physiological motion data with the Doppler radar system. Yet, it is very challenging to eliminate the noise associated with surroundings, especially with a single-channel Doppler radar system. However, single-channel Doppler radars provide the advantage of operating at lower power. Additionally, the heart rate extraction using single-channel Doppler radar has remained somewhat unexplored. This has motivated us to develop effective signal processing algorithms for signals received from single-channel Doppler radars. In this paper, we have proposed and studied three algorithms for estimating heart rate. The first algorithm is based on applying FFT on an FIR filtered signal. In the second algorithm, autocorrelation was performed on the filtered data. Thirdly, we used a peak finding algorithm in conjunction with a moving average preceded by a clipper to determine the heart rate. The results obtained were compared with the heart rate readings from a pulse oximeter. With a mean difference of 2.6 bpm, the heart rate from Doppler radar matched that from the pulse oximeter most frequently when the peak finding algorithm was used. The results obtained using autocorrelation and peak finding algorithm (with standard deviations of 2.6 bpm and 4.0 bpm) suggest that a single channel Doppler radar system can be a viable alternative to contact heart rate monitors in patients for whom contact measurements are not feasible.**

I. INTRODUCTION

Ubiquitous detection and monitoring of heart activity have the potential for early detection and diagnosis of cardiovascular disease. Heart rate monitoring is typically carried out by Electrocardiogram (ECG) tests which utilize electrodes to detect electrical impulse signals resulting from heartbeats. Other than being universally used, ECG tests are cheaper than echocardiography, computed tomography, and magnetic resonance imaging. However, the requirement of physical contact is often considered inconvenient due to the hassle of attaching multiple electrodes to the patient's body and making it unfavorable for skin sensitive or burn injury patients [1,2]. Consequently, these disadvantages have necessitated the need for non-contact remote monitoring of heart activities. In this case, microwave Doppler radar has proven to be a promising tool that allows the assessment of physiological parameters such as heart rate, heart rate variability, arterial pulse wave, and tidal volumes without affecting patients' physiological activity patterns [2,3]. Oh et. al. have presented an algorithm where the surrounding environment noise is acquired first using a low pass filter and high pass filter consecutively and is then subtracted step by step from the original signal. Afterward, the heart rate is detected through differential polarization with set threshold values. However, this algorithm is only applicable to signals from the quadrature (IQ) receiver Doppler radar system [1]. Although IQ architecture allows avoiding null demodulation points, its complexity and power consumption are both greater than the single-channel architecture [7]. Unlike Quadrature radar, a single-channel radar signal does not require demodulation which simplifies the signal processing steps. While subspace-based algorithms like multiple signal classification (MUSIC) have advantages of high resolutions, these algorithms but have a problem of very high complexity [8]. Thus, in this paper, we have proposed three different algorithms for heart rate detection using single-channel Continuous Wave (CW) Doppler radar. Our study works on extracting heart rate from a single target at a still motion. It will also be demonstrated that these three algorithms work with a fairly well resolution with reduced complexity. We have first investigated different filter cut-offs and established method of filtering the output signal. Then, we have applied three techniques of Fast Fourier Transform (FFT), autocorrelation, and peak finding algorithm to estimate heart rate. These approaches were applied to data collected with five different subjects. The underlying principle of a homodyne single-channel CW Doppler radar system is described in Section 2. Section 3 presents the experimental results obtained using the single-channel Doppler radar system and the paper concludes in Section 4.

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II. THEORETICAL BACKGROUND

Doppler theory states that a constant frequency signal reflected off an object with periodically varying displacement, $x(t)$ will undergo a time-varying phase shift while keeping the same frequency as it originally had. If we use the chest as the target, demodulating this phase shift will then give a signal directly proportional to the chest position. This displacement will contain information about movement due to heartbeat and respiration, from which heart rates and signatures can be determined [9,10]. Fig. 1 provides an illustration of a homodyne single-channel Doppler radar system. A single tone signal is transmitted from the CW at a frequency f [Hz] which can be represented as:

$$T(t) = \cos(2\pi f t) \qquad (1)$$

This signal is then reflected off the target at a distance d_o[m], with a time-varying displacement, $x(t)$ [m]. The distance and

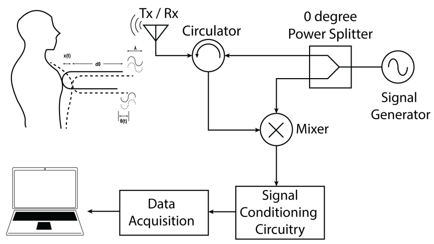

Fig. 1. Homodyne single-channel Doppler radar system for detecting physiological motions.

time delay between the transmitter and the target can be given as $d(t)$ and t_{do} respectively where,

$$d(t) = d_o + x(t) \tag{2}$$

$$t_{do} = \frac{d(t)}{c} \tag{3}$$

Since the actual distance between the antenna and the chest at the time of reflection can be expressed as $d(t - t_{do})$, the round trip time delay [s] is:

$$t_d = \frac{2d(t - \frac{d(t)}{c})}{c} = \frac{2}{c}d_o + x(t - \frac{d(t)}{c}) \tag{4}$$

Thus, a time-delayed signal with attenuated amplitude by a factor of A_R will be received at the receiver side. The received signal with a constant phase shift can be expressed as:

$$R(t) = A_R \cos\left(2\pi f(t - t_d)\right) \tag{5}$$

Substituting for the round trip delay, t_d the received signal becomes :

$$R(t) = A_R \cos\left[2\pi f t - \frac{4\pi}{\lambda}\left(d_o + x\left(t - \frac{d(t)}{c}\right)\right)\right] \tag{6}$$

Since a homodyne receiver is used in our architecture, the output signal is a baseband signal found by mixing the local oscillator signal with the received signal. Let, $L(t)$ is the time-varying local oscillator signal with negligible amplitude variation. Then $L(t)$ can be expressed as:

$$L(t) = \cos(2\pi f t) \tag{7}$$

The baseband signal with amplitude, A_B can be written as:

$$B(t) = A_B cos\left[\frac{4\pi}{\lambda}\left(d_o + x\left(t - \frac{d(t)}{c}\right)\right)\right] \tag{8}$$

The constant phase term, $\left(d_o + x\left(t - \frac{d(t)}{c}\right)\right)$ in the baseband signal determines null/optimum for a single-ended receiver. Further signal processing is done on this output baseband signal to extract useful information like heart rate.

III. EXPERIMENTAL METHODS AND OUTCOMES

The experiments for this study were conducted according to the Committee on Human Studies (CHS) protocol number 14884 which was approved by the CHS of the University of Hawaii system. We have used a homodyne single-channel radar with a patch antenna operating with a passband frequency in the 2.4 GHz ISM band. The sampling rate of the radar system employed in this study is set at 100 Hz. The radar is set at a distance of 1 m from the subject and data is recorded for approximately 3 minutes. For this experiment, the set of subjects consisted of five healthy adults with ages ranging from the middle twenties to the middle thirties. Datasets were collected from the subject in a still and seated condition for controlled respiration rates of 1) 12 breaths per min, 2) 15 breaths per min and 3) 18 breaths per min. These respiration rates were maintained by asking the subjects to breathe at the rate of a metronome application. The radar signals were processed using signal processing code written in MATLAB. Heart rates extracted from the data with the three algorithms were compared with the reference data taken using a pulse oximeter.

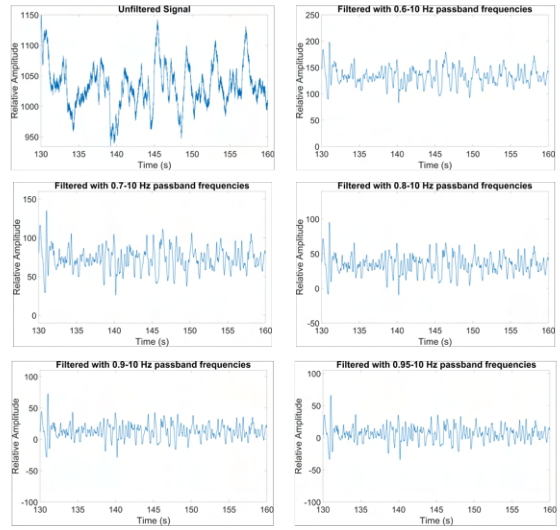

Fig. 2. Example of an unfiltered signal and the same signal filtered with 200-order FIR filter with high cutoff of 10 Hz and low cutoff (f_L) of 0.6 Hz - 0.95 Hz. The unfiltered signal is dominated by respiratory motion, and heart/pulse motion can be seen in the filtered signals. As $f_L$is increased, the amplitude of pulses is more consistent, which leads to more accurate heart rate extraction.

For filtering the noisy signal, a finite impulse response (FIR) bandpass filter with Kaiser window was used. The effect of lower cut-off frequency, f_L, of the filter on heart rate detection was studied for all subjects by changing it from 0.6 Hz to 0.95 Hz. As shown in Fig. 2, the signals with small amplitudes become more prominent when f_L is increased gradually. Even though the respiratory rates are low, there are harmonics that gets through if the cut off is set too low. Additionally, the heart rate found for 0.95 Hz was compliant with the reference value. Table I. lists the heart rates of the third subject obtained by varying f_L when the breathing rate is at 15 breaths per minute. It can be seen that the peak finding algorithm does not show heart rate for frequencies below 0.95 Hz.

Thus, the passband frequencies were set at 0.95 Hz to 10 Hz and the order of the filter was set as 200 [11]. Four different windows such as Hamming, Hanning, Blackmann-Harris, and Kaiser were used to find which one gives the best filter response. Among these, we have selected the Kaiser

TABLE I
DETECTED HEART RATES AT DIFFERENT LOWER CUT-OFF FREQUENCIES USING THREE ALGORITHMS.

f_L (Hz)	FFT	Autocorrelation	Peak Finding	Reference
0.6	60 bpm	72 bpm	No data	72 bpm
0.7	60 bpm	72 bpm	No data	72 bpm
0.8	60 bpm	72 bpm	No data	72 bpm
0.9	60 bpm	72 bpm	No data	72 bpm
0.95	87 bpm	70 bpm	70 bpm	72 bpm

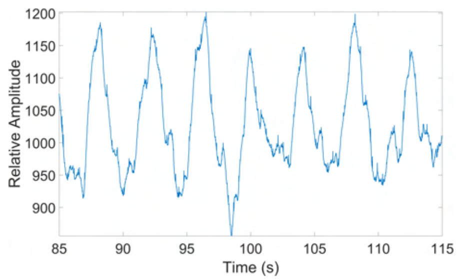

Fig. 3. Unfiltered signal from data taken at 15 breaths per min.

window with its shape factor beta set at 6.5 for the finest sidelobe attenuation. Fig. 3 illustrates the signal prior to the filtering is dominated by the respiratory signal where the heart signals are not visible. The same signal, bandpass filtered, is shown in Fig. 4. In this study, the heart rate of the subjects is extracted by performing Fast Fourier transform (FFT), autocorrelation, and peak finding methods. Before applying the rate finding algorithms the signal is filtered with 200 order FIR bandpass filter from 0.95 Hz to 10 Hz to isolate the heart signal. Then different approaches are applied. For FFT rate finding, the FFT is performed on the signal, and the peak frequency is used to calculate the heart rate. For the autocorrelation approach, the filtered signal is autocorrelated with lag from 1 to 200 samples, and the first local maxima is used to estimate the heart rate. Fig. 5-6 shows the extracted heart rate using these first two methods when the respiration rate is at 15 breaths per min. The third algorithm works by integrating a moving average filter with peak finding algorithm. At first, the average amplitude of all the positive sample is calculated and the value is set as a threshold. Any sample positive amplitude having value higher than the threshold is clipped. The same process is done for the samples with negative amplitude as well.

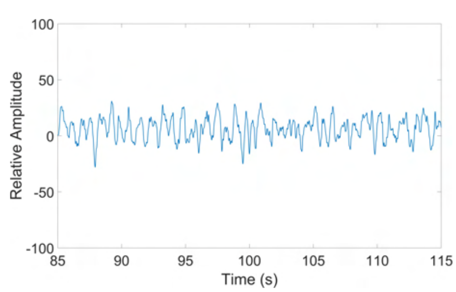

Fig. 4. Filtered signal using a 200th order FIR bandpass filter with 0.95 Hz -10 Hz cut off frequency from data taken at 15 breaths per min.

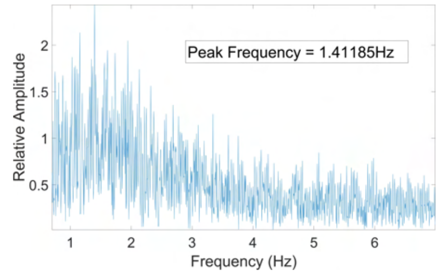

Fig. 5. Applying FFT to find the highest frequency representing the heart of the subject.

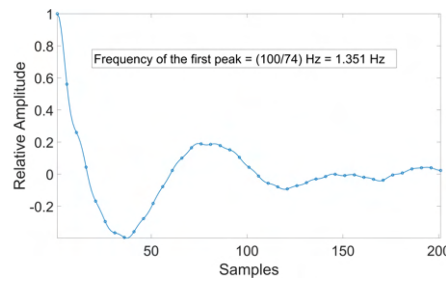

Fig. 6. Autocorrelating the filtered signal to find the frequency of the first peak corresponding to the heart rate of the subject.

This helps to get rid of the sudden undesirable spikes in the data. Next, the clipped signal is then passed through a moving average of 60 samples. The moving average is used to smooth the bandpass filtered signal [1]. Finally, peaks are identified by evaluating each sample to determine if it is the maximum when it is at the center of a 90-sample window. The heart rate is then found by dividing the total number of maxima by the overall sample time. Fig. 7 illustrates the peak finding algorithm used. The heart rates acquired from the different datasets and extracted using the mentioned algorithms are listed in Table II-IV. To evaluate the accuracy of these rate-finding algorithms, the difference between the average heart rate estimated with the Doppler radar system and the average heart rate from the refrence was calculated for each measurement. The mean and standard deviation of the absolute value of the difference is shown in Table V. The data indicate that the peak-finding method provided heart rates closest to those of the reference.

In comparison to studies employing complex signal processing like discrete-time wavelet transform for detection of QRS complex of ECG signal, our proposed algorithms use simpler techniques to extract heart rates from less defined received signal from Doppler radar with acceptable accuracy [12]. ECG signals contain baseline wander and powerline interference which need to be removed before signal processing. Thus, a complicated signal processing algorithm such as eigenvalue decomposition of Hankel matrix is required for filtering the ECG signal [13]. However, denoising our radar signals can be done easily with a windowed FIR filter which is computationally faster.

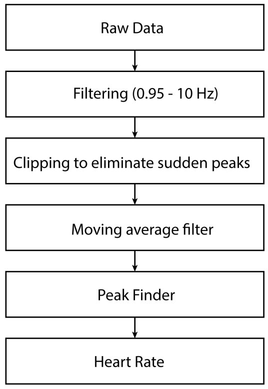

Fig. 7. Peak Finding Algorithm used for finding heart rate.

TABLE II
EXTRACTED HEART RATES FOR 12 BREATHS PER MIN USING THREE ALGORITHMS

Subject	FFT	Autocorrelation	Peak Finding	Reference
1	103 bpm	98 bpm	105 bpm	105 bpm
2	96 bpm	87 bpm	91 bpm	90 bpm
3	72 bpm	73 bpm	79 bpm	73 bpm
4	84 bpm	85 bpm	72 bpm	80 bpm
5	75 bpm	79 bpm	80 bpm	84 bpm

TABLE III
EXTRACTED HEART RATES FOR 15 BREATHS PER MIN USING THREE ALGORITHMS

Subject	FFT	Autocorrelation	Peak Finding	Reference
1	98 bpm	94 bpm	96 bpm	95 bpm
2	85 bpm	81 bpm	80 bpm	81 bpm
3	87 bpm	70 bpm	70 bpm	72 bpm
4	84 bpm	89 bpm	82 bpm	82 bpm
5	81 bpm	84 bpm	88 bpm	91 bpm

TABLE IV
EXTRACTED HEART RATES FOR 18 BREATHS PER MIN USING THREE ALGORITHMS

Subject	FFT	Autocorrelation	Peak Finding	Reference
1	104 bpm	98 bpm	104 bpm	105 bpm
2	82 bpm	90 bpm	88 bpm	92 bpm
3	68 bpm	67 bpm	62 bpm	69 bpm
4	82 bpm	79 bpm	79 bpm	79 bpm
5	81 bpm	82 bpm	88 bpm	87 bpm

TABLE V
ACCURACY OF THE THREE ALGORITHMS ASSESSED AS THE ABSOLUTE VALUE OF THE DIFFERENCE BETWEEN THE ALGORITHM OUTPUT AND THE REFERENCE RATE

Algorithm	Mean Difference	Standard Deviation
FFT	3.5 bpm	2.7 bpm
Autocorrelation	5.1 bpm	4.0 bpm
Peak Finding	2.6 bpm	2.6 bpm

IV. CONCLUSIONS

In this study, three algorithms were tested to find heart rate using a single-channel Doppler radar system. Based on the results, it can be stated that the non-contact method of heart rate measurement utilizing a single channel Doppler radar can be considered as an effective tool for heart rate monitoring and detection. While this worked assessed average heart rates over 3-minute samples, our future work will include algorithm development and assessment for beat-to-beat analysis on a larger number of subjects.

ACKNOWLEDGMENT

This work is supported in part by the National Science Foundation (NSF) under grants IIP-1831303 and CNS-2039089. Dr. Boric-Lubecke holds equity and serves as president of Adnoviv, Inc, a company that is the prime awardee of NSF STTR IIP-1831303 grant. The University of Hawaii has granted a license to Adnoviv, Inc. to commercialize Doppler radar technology for occupancy sensing purposes, and owns equity in Adnoviv, Inc.

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