Quadrature Demodulation with DC Cancellation for a Doppler Radar Motion Detector

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Abstract— **Continuous wave direct-conversion microwave Doppler radar can be used to detect cardiopulmonary activity at a distance. Demodulation sensitivity to a target's distance in a single channel system can be solved by using a quadrature direct-conversion receiver. A particular challenge in this technique is accurate demodulation in the presence of unwanted dc offset. A center tracking compensation method has been proposed to overcome this issue. In this paper, we present an estimation method for center tracking, and introduce a linear demodulation method. While non-linear (arctangent) demodulation is required for larger displacement or higher frequency systems, linear demodulation also provides accurate results for small displacement at lower frequencies at lower computational cost. In addition, we propose to use a real-time dc cancellation method, prior to center tracking compensation, to fully exploit the dynamic range of the pre-amplifiers and the resolution of the ADC, with minimum time delay and distortion. Experimental results demonstrate that linear and non-linear demodulation with the dc canceller result in accurate heart rate measurement with a standard deviation of less than 2 beats per minute. Large motion of the target, such as walking, was also successfully measured using non-linear demodulation.**

Index Terms—**Biomedical monitoring, biomedical signal detection, Doppler radar, dc offset, direct conversion quadrature receiver, linear demodulation, non-linear demodulation, eigen vector.**

I. INTRODUCTION

Quadrature direct-conversion microwave Doppler-radar has been introduced for non-contact detection or monitoring of human cardiopulmonary activity [1,2]. This method can be a promising tool for health care, emergency, military, or security applications, if reliable and robust sensing can be provided. One challenge in providing robust sensing is accurate demodulation of quadrature outputs, particularly in the presence of a large dc offset. It was previously demonstrated that a non-linear (arctangent) demodulation method with dc offset compensation can achieve a high degree of accuracy for heart signal extraction [6]. This method requires calibration while in the measurement environment and it cannot account for clutter dc caused by subject's stationary body parts which can cause additional dc offset. A center tracking compensation method was proposed in [7] to overcome this issue by estimating dc information from the quadrature output signals. In this paper, we present an estimation method for center tracking, and introduce a linear demodulation method. While non-linear demodulation is required for larger displacement [7] or higher frequency systems, linear demodulation also provides accurate results for small displacement at lower frequencies. In addition, we propose to use a real-time dc cancellation method [8], prior to center tracking compensation, to utilize the full dynamic range of the pre-amplifiers and maximum resolution of the ADC, with minimum time delay and distortion. The mathematical derivation of the center tracking estimation method is presented, including steps that lead to the linear demodulation method. Linear and non-linear demodulation methods are compared theoretically, indicating that linear demodulation is as accurate as non-linear demodulation of cardiopulmonary motion for phase deviation of less than $46.8°$, corresponding to a displacement of 0.81 cm at 2.4 GHz. Experimental results demonstrating that center tracking is suitable for demodulation of small and large displacement are presented. In addition, it is shown that at 2.4 GHz linear and non-linear demodulation result in similar heart signal detection accuracy with some small differences, confirming the theoretical prediction.

II. CENTER ESTIMATION AND DEMODULATION

In this section cardio-pulmonary quadrature signal properties are discussed, and the mathematical derivation of center estimation is presented. Linear demodulation is introduced as one of the steps required for center estimation. Linear and non-linear demodulation methods are also compared as a function of phase deviation.

Typically, a Doppler radar motion sensing transceiver transmits a radio wave signal and receives a phase-modulated signal reflected from a target. A block diagram of a quadrature Doppler transceiver is shown in Fig. 1.

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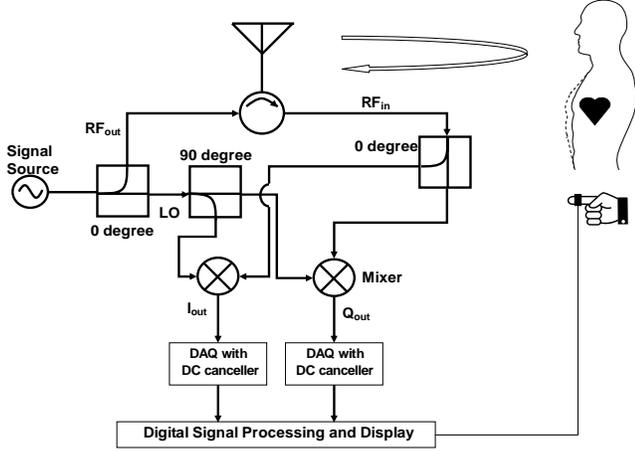
Fig. 1. Block diagram of a quadrature Doppler radar system with dc canceller. The LO signal is divided by a two-way 90° power splitter to get two orthonormal baseband signals (I and Q) which are combined to demodulate the subject's motion.. Demodulated output results are compared with a wired finger pulse sensor reference.

A single source supplies both the RF output and LO signals. The LO signal is further divided using a 90° power splitter to provide two orthonormal baseband outputs.

Assuming the target's motion variation is given by $\Delta x(t)$, the quadrature baseband output, assuming balanced channels, can be expressed as:

$$B(t) = A_r \exp\left[\theta + 4\pi\Delta x(t)/\lambda\right], \qquad (1)$$

where θ is the constant phase shift related to the phase change at the surface of a target and the phase delay between the mixer and antenna. Applying arctangent demodulation [6] to the ratio of the quadrature outputs, phase information linearly proportional to target's motion can be extracted as:

$$\phi(t) = \arctan\left(\frac{B_Q(t)}{B_I(t)}\right) = \arctan\left(\frac{A_r \sin(\theta + p(t))}{A_r \cos(\theta + p(t))}\right) = \theta + p(t), \quad (2)$$

where $p(t) = 4\pi\Delta x(t)/\lambda$ is the superposition of the phase information due to the target's motion. However, dc offsets in the quadrature channels act as a linear transform on the I and Q components, and thus arctangent demodulation output becomes:

$$\phi'(t) = \arctan\left(\frac{B_Q(t)}{B_I(t)}\right) = \arctan\left(\frac{V_Q + A_r \sin(\theta + 4\pi\Delta x(t)/\lambda)}{V_I + A_r \cos(\theta + 4\pi\Delta x(t)/\lambda)}\right), \quad (3)$$

where V_I and V_Q refer to the dc offsets of each channel resulting from the finite port to port isolation of the transceiver as well as from clutter reflections. The dc signal contains this dc offset as well as the dc information associated with target's position required for accurate demodulation. As shown in Fig.

2, when there is only one source of periodic motion, the complex plot of quadrature outputs forms a fraction of a circle that has a radius of signal amplitude, A_r, with its center offset by the dc offset of each channel. This property allows proper elimination of dc offset and preservation of all the desired information including ac and dc signals associated with target's periodic motion, with the latter being the magnitude of the radius projected on each axis when the center of the arc formed by target motion is tracked back to the origin of the complex plot. We have proposed this "center tracking" technique in [7].
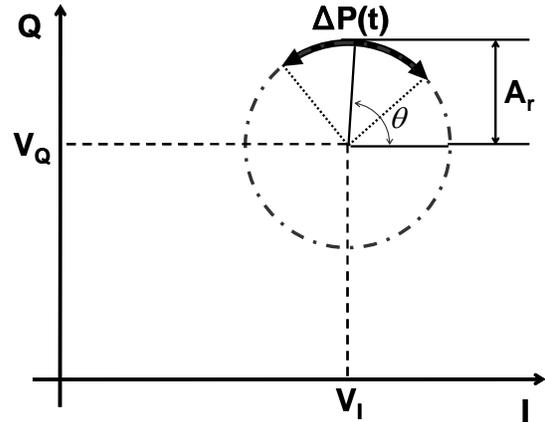

Fig. 2. Complex constellation of quadrature outputs due to the target's periodic motion. $\Delta P(t)$. V_I and V_Q are the dc offsets of I and Q channels respectively, and A_r^2 is the received signal power.

The complex plot of the I and Q outputs depends both on received signal power and phase deviation due to a target's motion. From (1) received signal power becomes A_r^2, the square root of which is the radius of the arc formed by phase deviation from a target's motion. Phase variation, which is proportional to the arc length, is proportional to the ratio of target's motion over wavelength of the carrier signal. In other words, arc length becomes longer either due to an increase in the target's actual motion or due to an increase in the carrier frequency. Consequently, when a target is moving with a large deviation resulting in changing received signal power, the radius of the arc will vary while its center remains located at the same point, thus forming a spiral-like shape rather then a circle. On the other hand, when operating frequency increases so that the small physical motion of a target is converted to a large phase variation, a longer arc length on the circle results.

After sampling, the received baseband signal in (1) can be written as:

$$B[n] = A_r \exp(j(p[n] + \theta)) + k + w[n], \qquad (4)$$

where $w[n]$ is the system noise, which is assumed to be identical independently distributed (iid) circular Gaussian, and

k is a complex value representing the dc offsets due to clutter noise or system device imperfection.

In order to do arctangent demodulation, the dc offset must be estimated. This is done in two steps [9]: first, the arc is rotated so that it is parallel to the Q-axis, i.e., with the angle -θ. Then we find the center of the circle, i.e., k, which is now located on the I-axis, that is k is real.

First the data is multiplied by the transpose of the matrix of eigenvectors of the covariance matrix to rotate the arc. Thereby the dc offset becomes an offset purely on the I-axis. Since after rotation, the Q-component is always in an optimum point (and the I-component always in a null point), we can simply use the Q-component as the demodulated signal. In the following we will denote this *linear demodulation*.

The next step after rotation is estimation of *k*. We have developed a heuristic estimator.

$$\hat{k}(n_1, n_2) = \frac{|B[n_1]|^2 - |B[n_2]|^2}{2\operatorname{Re}\{B[n_1] - B[n_2]\}}, \qquad (10)$$

$$\hat{k} = \operatorname*{median}_{n_1 \neq n_2}\{\hat{k}(n_1, n_2)\}. \qquad (11)$$

The performance of different demodulation methods depends on the ratio A_r^2/σ^2, where σ^2 is system noise power, (which could be considered a "passband SNR") and the maximum arc length θ_m. As can be seen in Fig. 3, for low arc length, linear demodulation is better than non-linear, however non-linear demodulation becomes better at θ_m=0.13. This arc length corresponds to a displacement of 0.81 cm at 2.4 GHz.
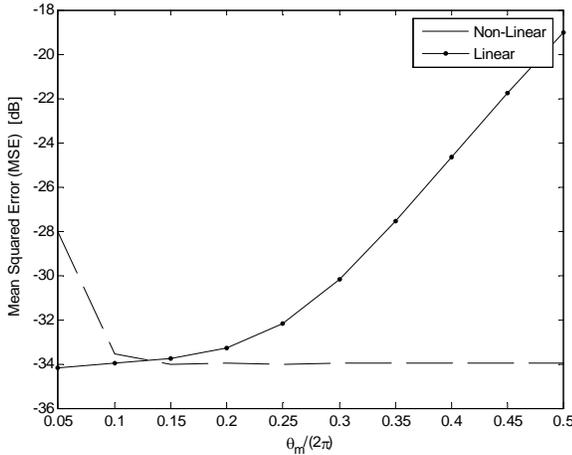

Fig. 3. Performance of different demodulation methods for A$_r$=100, k=50, σ^2 = 2.

III. DATA ACQUISITION WITH DC CANCELLER

The purpose of the dc canceller is to cleanly remove the dc components of I and Q outputs (Fig. 1) with minimum time delay and distortion. The remaining time-varying signal can thus be amplified and sampled with maximum resolution. The proposed data acquisition system with dc canceller is composed of two stages as shown in Fig. 4. The first stage includes a 16-bit analog-to-digital converter (ADC1) and a 16-bit digital-to-analog converter (DAC). This stage acquires an estimated value of the dc offset and provides the reference level for the second stage. The second stage removes the dc offset through differential amplification and sends the remaining time-varying signal to a second 16-bit ADC (ADC2) for quantization.
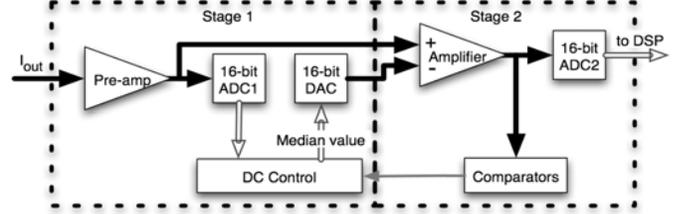

Fig. 4. Data acquisition system with the dc canceller. The first stage acquires an estimated value of the dc offset and provides the reference level for the second stage. The second stage removes the dc offset through differential amplification and sends the remaining time-varying signal to a second 16-bit ADC (ADC2) for quantization.

The input to the first signal stage includes a large dc offset as well as the small ac signal that contains physiological motion information. A fixed gain pre-amplifier (SRS560) with 10 Hz bandwidth is used for anti-aliasing, and to provide proper signal amplitude from the RF mixer to the first 16-bit ADC (NI 6259), ADC1. At the start of the acquisition cycle, ADC1 instantly acquires a value from the signal. This value is the initial estimated dc offset. This initial value is given to the DAC (NI 6259) and the DAC outputs the initial estimated dc offset as a reference voltage level. In essence, the ADC-DAC pair acts as a sample-and-hold (S/H) unit. The dc control unit is a LabView implemented code interfaced through the data acquisition board (NI 6259). To ensure that the value provided to the DAC maximizes the dynamic range of the data acquisition system, the dc control unit provides correction to the dc reference. Baseline drift of the signal due to slight changes in subject position or temperature will eventually require a correction in the dc reference value. A mid-point value is the best value to optimize the range of a periodic signal. Because physiological signals associated with pulmonary activity are of low frequency, several seconds are needed before obtaining a new dc reference level. To find the midpoint of the minimum and maximum of a respiration period, 6 seconds of initial data was used to estimate the position related dc value. This time period can be lowered to 2-3 seconds, as needed. For comparison, if an SRS560 pre-amp was used with ac coupling or in single-pole high-pass mode with a 0.03 Hz cut-off, the settling time would be on the order of 30 seconds. In addition, a traditional S/H is susceptible to voltage droop due to transistor parasitic resistances that will drain the charge away from a holding capacitor. Utilizing an ADC-DAC pair instead, effectively provides a zero-droop S/H.

The second stage subtracts the estimated dc offset from the DAC, from the input signal using a differential amplifier (SRS560). The signal is then sampled by ADC2 (NI 6259). LabView software analyzes the data in which comparators check for near-clipping conditions of the amplified signal. A near clipping condition could be a signal that exceeds 95% of the full-range, causing the dc control unit to update the DAC to

the last available value from ADC1. In order to maximize the system dynamic range before acquisition of the physiological signals by the ADC2, a system calibration to compensate for additional dc offsets introduced by the DAC and amplification stage is implemented. This is achieved by an initial calibration cycle before data acquisition or by user input.

IV. EXPERIMENTAL RESULTS

The coaxial quadrature radar system and measurement set-up with dc canceller used for these experiments is shown in Fig. 1. Data is collected from a seated subject facing the antenna at a distance of about 1 meter, and from a moving subject walking back and forth with 150 cm deviation from an initial distance of 100 cm. A commercially available Antenna Specialists ASPPT2988 2.4 GHz patch antenna was used, with a gain of 7.5dBi, an E-plane range of 65°, and an H-plane range of 80°. An HP E4433B signal generator was also used, divided into RF and LO signals by a Mini-Circuits ZFSC-2-2500 signal splitter. A Narda 4923 circulator was used to isolate transmit and receive signals, with a measured RF to LO isolation of -22 dB. The LO signal was further divided by a hybrid splitter, Narda 4033C, to provide quadrature outputs. A Mini-Circuits ZFM-4212 was used for the mixer in each channel. Mixer outputs were filtered from dc to 10 Hz for anti-aliasing by an SRS560 LNA (Pre-amp in Fig. 4) and digitized at 100 Hz sampling rate. Six hundred samples, or 6 seconds of digitized signals were processed to get the mid-point or dc values for the signals. These mean values were then subtracted from the filtered signals, and the remaining ac signals amplified by differential amplifiers.

The resulting amplified ac signals, sampled at a 100 Hz rate, were further processed in the digital domain. First ten seconds, or 1000 samples of data were used for estimating the center of arc, corresponding to more than one cycle of respiration and thus forming a sufficient arc length for center tracking. Linear demodulation was performed as one of the steps required for center tracking. Quadrature signals that form an arc centered at the origin in the complex plot were combined using non-linear demodulation. Demodulated outputs were digitally filtered by an FIR filter with a frequency range of 0.8 to 10 Hz to obtain heart signal. Heart rates were then extracted in near-real time with custom software based on an autocorrelation algorithm described in [10] a with window size of 4 seconds, and compared with rates obtained from a wired finger pressure pulse sensor (UFI 1010) used as a reference.

A large arc length in phase tracking can result either from using a higher operating frequency or measuring physically larger motion. The latter case was employed in this experiment, through tracking a walking subject's movement. The same arctangent demodulation method explained above can be used for demodulation. However in this case since the phase variation caused by the target's motion is much bigger than 2π or a half wavelength, (6.25 cm at 2.4 GHz), the arctangent demodulated output needed to be unwrapped as the complex plot was no longer a small fraction of a circle but spiral-like in shape, with the same center point. This occurs because the dc offset caused by clutter or leakage within the device is fixed, while the received signal power, which corresponds to the radius of the complex signal circle, varies associated with a target's distance from the antenna. In contrast, in the case of using a higher frequency radar system with small physical motion, only a phase deviation increase without a decrease in received power would be expected.

Figures 5 and 6 show the I, Q, linear, and non-linear demodulated signals obtained using the measurement setup shown in Fig. 1, for a subject in an intermediate position close to a null position for the Q channel (Fig. 5), and close to a null position for the I channel (Fig. 6). The null and optimum positions cannot be set exactly for heart rate measurements, as the nominal distance (and associated phase) varies as a result of respiration and affects heart rate data accordingly. The different arc lengths in Fig. 5 (a) and Fig. 6 (a) result from different respiration patterns. The data in Fig. 5 was obtained with the more shallow breathing for the subject. The arc formed by respiration motion of the chest is first rotated to be parallel with the Q axis, and then tracked back to the origin using center estimation (Fig. 5 (a) and 6 (a)). The rotated signal is used for linear demodulation, and the center-tracked signal is used for non-linear demodulation. Since the measurements in Fig. 6 were taken for the subject in a position close to the optimum on Q channel, in this case rotated and center-tracked signals almost overlap. To examine the effectiveness of linear and non-linear demodulation, standard deviation was used to provide a quantitative comparison of accuracy. As shown in Figures 5 and 6, a drop-out region, which is defined as the region where BPM differs from reference by more than 30, occurs at the null point due to either degradation in signal power or phase alternation, and this region was excluded when calculating standard deviation. In Fig. 5 (a), the arc length is about 43 degrees, corresponding to θ of 0.12 radians. In this case, the I and Q channel outputs show an error of 1.7 and 4.9 beats, respectively, during the 65 second time interval while the linear and non-linear demodulation have an error of only 1.5 or 1.6 beats, respectively. In Fig. 6 (a), the arc length is about 75 degrees, corresponding to θ of 0.21 radians. In this case the I and Q channels show an error of 7.1 and 2.0 beats, respectively, during the 65 second time interval while the linear and non-linear demodulation have an error of only 2.3 or 1.8 beats, respectively. These experimental results confirm that linear demodulation yields better accuracy for arc lengths of less than 0.13 radians, while non-linear demodulation performs better for larger arc lengths. While both linear and non-linear demodulation can be used for tracking of small motion at moderate frequencies, in the case of large motion such as walking, non-linear demodulation is required for accurate recovery of displacement.

Fig. 7 shows the I and Q output signals (a), complex plot of I and Q outputs after center tracking (b), and non-linear demodulation output for the case of a subject walking. For this measurement, the subject walked back and forth within a 150 cm distance, aligned with the antenna beam. As expected, the complex plot forms a spiral-like shape, due to the received
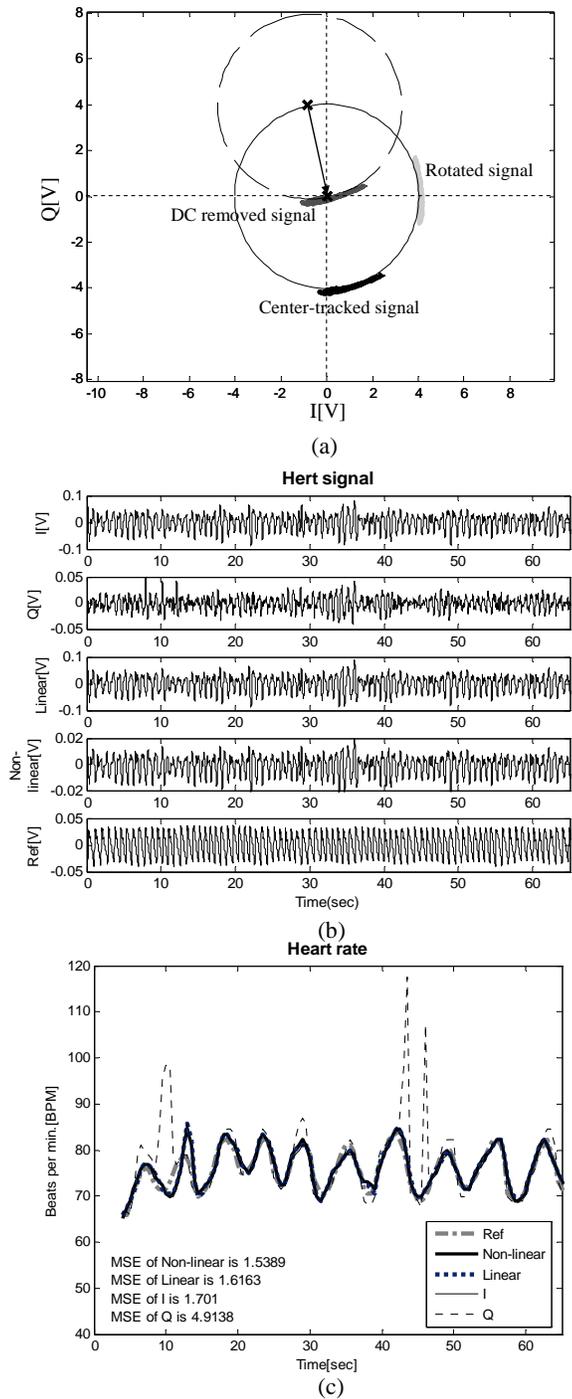
to distance using a $\lambda / 4\pi$ multiplication, or about a factor of 1, as indicated by (3). As shown is Fig. 7 (c), the calculated subject displacement was 150 cm as expected. From the measurement results described above it is evident that both linear and non-linear demodulation results are more accurate for a stationary subject than either of the single channel output signals, or at least as good as the signal from the optimum channel. Thus, linear and non-linear demodulation produce robust and accurate data for rate tracking regardless of the target's position, without need for channel selection. In addition, non-linear demodulation can be also used on signals with larger phase deviation resulting
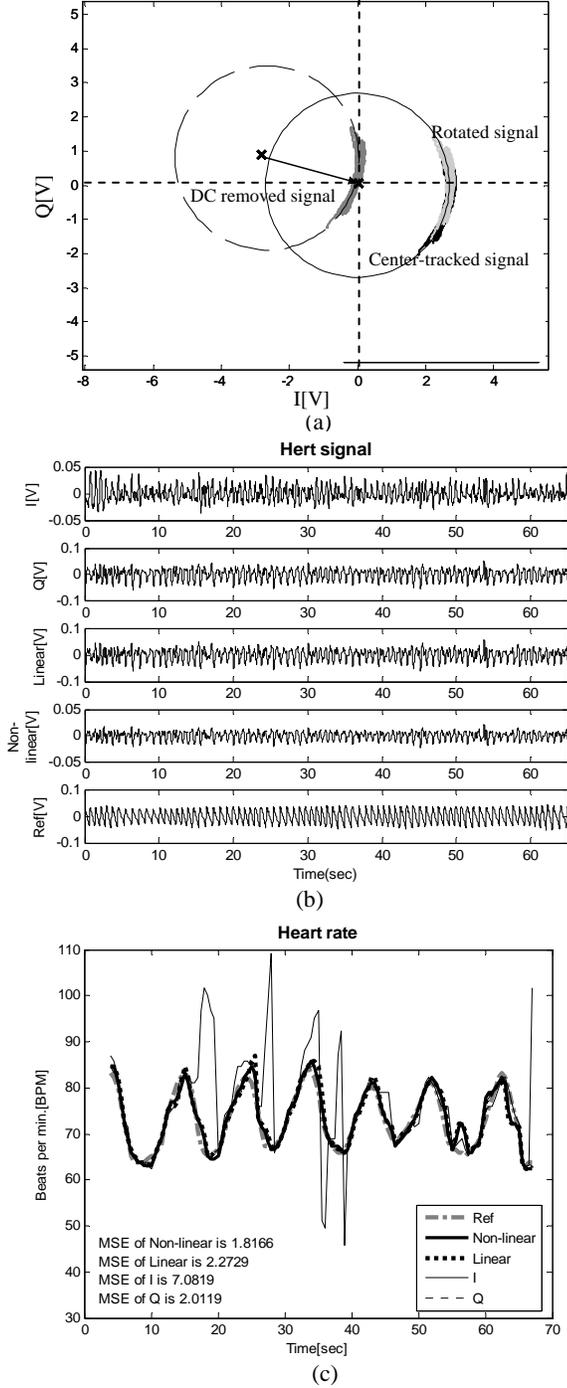

(a)

(b)

(c)

Fig. 5. I, Q, linear and non-linear demodulated signals measured for a position where the Q channel is close to a null condition. Arc formed by respiration motion of the chest is tracked back to the origin using center estimation (a). Doppler radar heart signals are extracted from I and Q channels, using linear and non-linear demodulation, and a reference heart signal obtained from finger pulse is shown for reference (b). Excluding drop-outs, defined when Doppler heart rate differs from the reference by more than 30 beats per minute, the I and Q channel data has an error of 1.7 or 4.9 beats, respectively, over the same 65-second interval where the linear and non-linear demodulation have an error of only 1.5 or 1.6 beats.

signal power variation (Fig. 7 (b)). The non-linear demodulation output, which is phase information linearly proportional to the actual distance variation, can be converted

(a)

(b)

(c)

(a)

(b)

(c)

Fig. 7. I and Q outputs (a), complex plot (b), and extracted displacement (c) for a walking subject.. Since the displacement of 150cm is much larger than the wavelength at 2.4 GHz, baseband I and Q outputs are frequency modulated according to speed of the target as well as amplitude modulated due to the receiving signal power variation (a). Complex plot of I and Q outputs forms complete circle with different radius but same center point, and dc offset is removed by dc canceller (b). Arctangent demodulation output can restore actual movement of a target by simply unwrapping output to compensate 2π ambiguity (c).

Fig. 6. I, Q, linear and non-linear demodulated signals measured for a position where the I channel is close to a null condition. Arc is formed by respiration motion of the chest (a). Digitally band pass filtered data extracts heart signal from a raw data (b). The I channel rate data (c) shows drop-out regions when the SNR is insufficient for digitization or phase alternation that happens when the chest position across angular position of zero. Excluding drop-outs, the I and Q channel data has an error of 7.1 or 2.0 beats, respectively, over the same 65-second interval where the linear or arctangent output has an error of only 2.3 or 1.8 beats.

from the use of higher frequencies or larger subject movement such as walking as shown in Fig. 7.
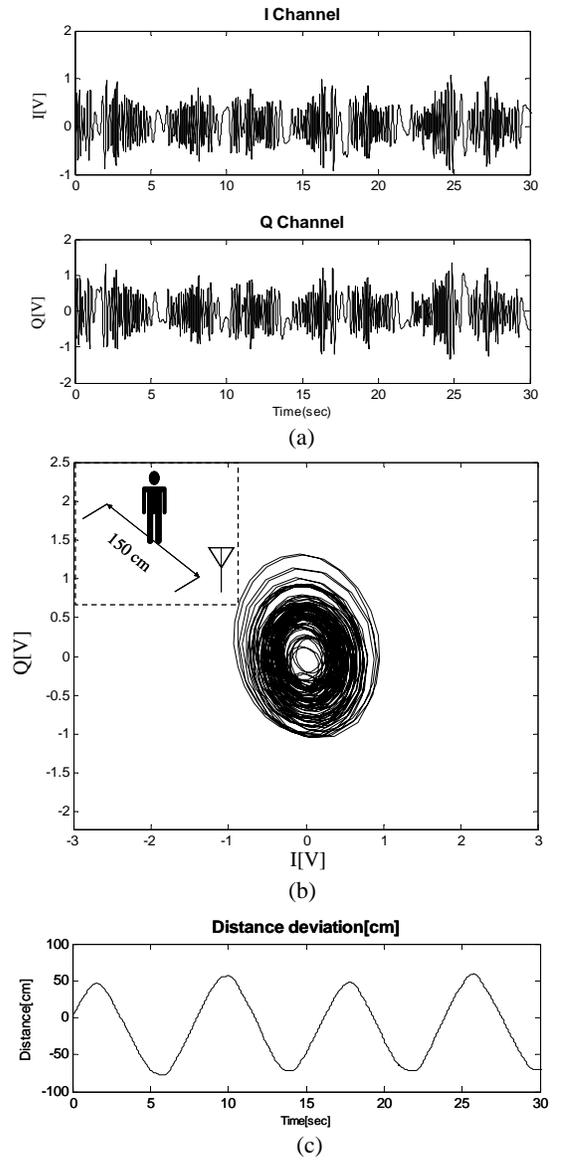
V. CONCLUSION

Non-linear quadrature demodulation has been proposed to overcome single channel Doppler radar limitations for cardiopulmonary sensing [6]. However, accurate dc compensation for dc offset resulting from hardware imperfections and clutter reflections is critical for performing non-linear demodulation. Dc cancellation with the described center tracking method provides a simple means of dc offset compensation. It was demonstrated that this method restores dc information signal directly from the I and Q signals associated with target's motion, while minimizing time delay and

distortion, in contrast with analog high-pass filter methods. Linear demodulation was also introduced as one of the steps required for center estimation, and the performance of linear demodulation was compared with non-linear demodulation, both theoretically and experimentally. Experimental results at 2.4 GHz demonstrated that both linear and non-linear demodulation with dc cancellation result in accurate heart rate measurement with a standard deviation of less than 2 beats per minute. While linear demodulation provides slightly higher accuracy for a small phase deviation, non-linear demodulation performs better for larger phase deviation, and is the only choice when large subject motion, such as walking, is considered.

VI. Acknowledgment

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