UWB RADAR SIGNAL PROCESSING IN MEASUREMENT OF HEARTBEAT FEATURES

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ABSTRACT

In this paper, a new signal processing algorithm in ultra wideband (UWB) radar as a non-contact sensor with the ability to precisely monitor heartbeat and respiration features is introduced. Exploiting harmonic series representation model that fits completely the return signal from a human chest-wall and wisely engaging an innovative time-varying filter, we propose a new free interference signal that indicates heartbeat features. Using this signal, a novel processing procedure in extraction of heartbeat rate (HR) and heart rate variability (HRV) is proposed. By experimental data, we investigate the applicability of the new approach for real scenarios comparing to other recently proposed procedure in different aspects.

Index Terms— *UWB radar, non-contact measurement, time-varying filter, heart rate variability.*

1. INTRODUCTION

All time monitoring of vital signs like breathing and heartbeat rate during the treatment process is considered as a critical factor in medical science. Nowadays, remotely sensors are an attractive alternative to electrode-attached sensors such as electrocardiography (ECG) since they monitor these signs without interfering the users activities. Many researches, conducted into the development of remote sensing systems, use microwave radars as the most papular approach [1]. Heart rate variability (HRV), mentioned to the beat-to-beat variation in heartbeat rate (HR), is an important indicator used in detection of some cardiac disorders such as diagnosis of stress syndromes and sudden cardiac death. Assessment of this indicator by radar has been sought in some pervious works. In these works, extraction of HRV is done by tracking the trace of extracted HR in time. Therefore, measurements of HR must be such accurate that the variation of rates can be tracked correctly.

Continuous wave radar known as Doppler radar is the prevalent idea in extraction of the HR and HRV. In these radars, the chest wall movement, made by respiratory and heartbeat motions, causes a Doppler phase modulation in the reflected signal. This Doppler effect as an exclusive effect of cardio-respiratory activities on the return signal is used in estimation of respiratory and heartbeat rate. Some of the main drawbacks of Doppler radars are the dependency of measurement accuracy to distance between radar and target, stationary clutter and crosstalk [2]. More recently, UWB impulse radars by their ability to extremely alleviate mentioned challenges are presented as a serious competitor Doppler radars in advanced studies [2–4]. This capability that stems mainly from the advantages of UWB radars such as high range-resolution and immunity to external interference, is obtained by paying a penalty which is noticeably low sensitivity to small movements such as heartbeat motions that makes problem in detection of HR and HRV. Therefore, the extension of these recent studies are limited in two domains: first, only focus on respiratory analysis [3] and second, estimating the phase of received signal contaminated by strong interferences and trying to relieve interferences [2, 5]. Because of high range-resolution in UWB radars, breathing motions are sensed in the time delay of the received pulse scattered from the chest wall. As a result, in UWB radar, cardio-respiratory activities, in addition to the Doppler phase modulation, are reflected in this time-varying delay. This is another advantage of UWB impulse radars over Doppler radar that is missed in all the advanced UWB radar studies. It means up to now, according to our knowledge, all UWB approaches only use one of time-varying delay [5, 7–9] or Doppler phase modulation [2] in extraction of vital signs. In this paper by considering both time-varying delay and Doppler phase modulation, the sensitivity is improved. Also, by exploiting a proposed time-varying filter and wisely using the derivation of received signal, a new interference free indicator is introduced. All these innovations are offered in the form of a processing algorithm to find HR accurately. Moreover demonstrating of superiority of the proposed algorithm over newly presented algorithm known as harmonic path algorithm (HAPA) [5] in different aspects, we show that this HR estimation is so accurate that the determination of HRV is possible by tracking of time trace of HR.

2. SYSTEM MODEL AND PROBLEM STATEMENT

In the considered system, a baseband narrow pulse waveform $u(t)$ with the repetition interval of T_p is modulated to operation frequency f_c and transmitted towards the chest wall of a human subject located at the distance of R from the radar. This transmitted signal in baseband can be expressed as:

$$s(t) = \sum_{n} u(t - nT_p). \tag{1}$$

Generalized Gaussian pulse (GGP) is usually used as a physically realizable narrow pulse waveform in transmitted signal and its time domain representation is:

$$u(t) = A \exp\left(-4\pi\left(\frac{t-t_0}{\Delta T}\right)^2\right), \quad 0 \leqslant t \leqslant T_p, \tag{2}$$

where A is the peak of waveform at time $t = t_0$; ΔT is the pulse width and $\Delta T + t_0 \ll T_p$.

When the transmitted signal $s(t)$ scatters back after impinging the chest wall and static objects in the vicinity of the subject, we can model the time-varying impulse response of channel as [6]:

$$h(t) = a_r \delta(t - \tau_r(t)) + \sum_{i=1, i\neq r}^{P} a_i \delta(t - \tau_i). \tag{3}$$

The second term of the right side of (3) is related to the static scatterers where $\delta(.)$ is the delta dirac; P is the total number of scatter

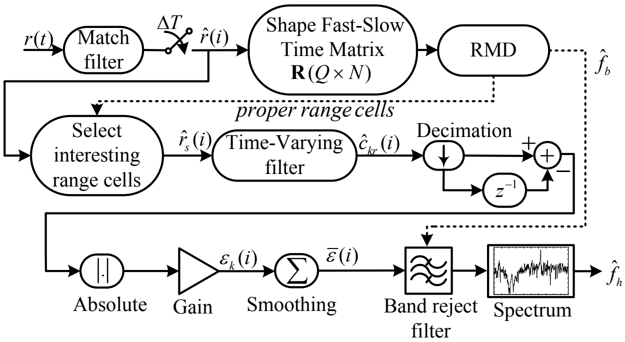


Fig. 1: Block diagram of novel signal processing algorithm in UWB radar.

Table 1: RMD procedure in detection of breathing rate.

1. Linear trend removal is subsequently subtracted from fast-slow time matrix, resulting in:

$$\hat{\mathbf{R}} = \mathbf{R}^T - \mathbf{x}\left(\mathbf{x}^T\mathbf{x}\right)^{-1}\mathbf{x}^T.\mathbf{R}^T$$

where $\mathbf{x} = [\mathbf{q}/Q \quad \mathbf{1_Q}]$, $\mathbf{q} = [0, ..., Q-1]^T$ *and* $\mathbf{1_Q}$ *is a* $Q \times 1$ *vector with unit values*

2. DFT over columns of clutter removed matrix $\hat{\mathbf{R}}$, resulting in:

$$\hat{\mathcal{R}}[\kappa, n] = \mathrm{DFT}\left\{\hat{\mathbf{R}}[q, n]\right\}$$

3. Frequency windowing on each slow-time dimension:

$$\widehat{R_w}[\kappa, n] = \widehat{R}[\kappa, n] \odot w[\kappa]$$

where $\odot$ *denotes pointwise multiplication and* $w[\kappa]$ *is the rectangular window that covers frequency components that may present breathing rate.*

4. Thresholding and detection.

points with the distance of R_i from the radar; $\tau_i = 2R_i/c$ (c is light speed) is the time delay of the received signal from i^{th} scatter point and complex coefficient a_i reflects the combination effects of transmitter and receiver antennas and radar cross section of the i^{th} scatter point. The first term of the right side takes into account the effect of scattering in the chest wall position. Because of breathing and heartbeat effects, the chest wall has a periodically movement. This motion causes time-varying delay for the received signal at this position. This time-varying delay is accurately and analytically modeled that is described by the following expression [6]:

$$\tau_r(t) = \frac{2R}{c} + \tau_b \sin(2\pi f_b t) + \tau_h \sin(2\pi f_h t), \tag{4}$$

where f_b and f_h are breathing and heartbeat rate respectively that $f_b \ll f_h$ in lifelike scenarios. The time-varying phase made by this delay, $\phi_r(t) = 2\pi f_c \tau_r(t)$, is known as Doppler phase modulation. The variation of $\tau_r(t)$ comparing to τ_i is so slow that the system assumed for channel model (i.e. (3)) can be considered time-invariant. So, the received signal in the baseband admits the following equation:

$$r(t) = a_r e^{j\phi_r(t)} \sum_n u(t - \tau_r(t) - nT_p) \tag{5}$$
$$+ \sum_{i=1, i\neq r}^{P} \sum_n a_i u(t - \tau_i - nT_p) + n(t),$$

where $n(t)$ is the AWG thermal noise. According to (5) the information of cardiopulmonary is hidden in the variation of the time delay of received signal from the chest position and also in the Doppler phase modulation, $\phi_r(t)$. The Doppler phase modulation that can be detected by energy detection in the spectrum of the received signal is the combination of respiratory and heartbeat rate. Actually, since $\tau_b \gg \tau_h$ in real world, respiratory rate masks the information of heartbeat rate. Also, there exist lots of interferences, effected by respiratory motion, in the spectrum of the received signal besides the respiratory and heartbeat rate [5]. This challenge makes accurate measurement of heartbeat rate impossible. This problem can be extremely relieved by two solutions. First, detecting the breathing rate and filter out respiratory motions from the Doppler phase modulation. In the long time observation of $r(t)$, a harmonic variation that represents breathing rate is sensed in the time delay. This rate can be detected from the harmonic variation by the algorithm known as respiratory-motion detection (RMD) [8]. By detecting breathing rate using RMD and filtering the portion of breathing activities in Doppler phase modulation, heartbeat term can be amplified. Second, since $f_b \ll f_h$, by derivation of $r(t)$ respect to t the heartbeat term to breathing term ratio is extremely amplified. Also this derivation removes all interferences and this challenge may be solved. However, because of wideband signal and noise in this system, derivation highly intensifies noise too. In the novel algorithm, presented in the next section, moreover using first solution, a novel procedure is used that boosts heartbeat term by differentiation while reduces the noise.

3. HEARTBEAT EXTRACTION ALGORITHM

3.1. Finding breathing rate & Removing clutters

In equation (5), since the τ with any indices and T_p are the order of nanoseconds, the time variable t as the argument of function $u(.)$ represents a variable that is famous as fast-time variable. Also, since f_b and f_h are small values, the time variation of $\tau_r(t)$ is sensed if it is observed during the time that is in order of seconds. Therefore, the time index t in $\tau_r(t)$ is famous as slow-time variable. In the presented algorithm, shown in Fig. 1, first conventional radar prototype is used to shape a matrix called fast-slow time matrix. The received signal is filtered by a filter that is matched to the $u(t)$ and has the time width of ΔT. The output of filter is a sampled signal, $\hat{r}(i)$ with sampling rate ΔT, that contains N samples during each period. Each sample indicates an especial range interval and is called range cell. L periods of sampled signal are averaged to smooth the AWG noise and make a range profile. Therefore, in each $T_{st} = LT_p$ seconds, one range profile is made. All N range cells in q^{th} (where $q = 1, ..., Q$) range profile shape the q^{th} row (fast dimension) of fast-slow time $Q \times N$ matrix $\mathbf{R}$. By considering 1) linear trend removal for removing stationary clutter; 2) using DFT over columns (slow dimension) of the clutter removed fast-slow time matrix; 3) frequency windowing of slow-time dimension; and 4) cell-averaging, empirically threshold setting and detection [9], all presented in RMD algorithm [8, 9], the interesting range cells where the chest is placed in and also the rate of breathing are detected. The RMD procedure is depicted in table 1. By detection of interesting range cells, the effect of all other static objects, presented as clutter, can be removed.

3.2. HR extraction

The transmitted signal $s(t)$ is a periodic signal and therefore, based on Fourier principle, it has a harmonic series representation as follows:

$$s(t) = \sum_{k=-K}^{K} b_k e^{j\frac{2k\pi}{T_p}t}, \tag{6}$$

where b_k are complex coefficients of Fourier series of $u(t)$; K shows the number of effective components and it is determined such that the amplitude of c_K is at least $\sqrt{2}$ times smaller than the dc component. As mentioned, in RMD procedure the static clutter in the interesting range cells are removed in linear trend removal process. Also, by finding the interesting range cells, other range cells and the effect of other objects can be deleted. So, the second component of the right side of (5) is not considered in next coming processes. Considering (6) and focusing on the effect of chest wall in (5), the received sampled signal after match filter, $\hat{r}_s(i)$, can be restated as:

$$\hat{r}_s(i) = \sum_{k=-K}^{K} c_{kr}(i)\, e^{j\frac{2k\pi}{T_p}i} + \hat{n}(i) \qquad (7)$$
$$c_{kr}(i) = \alpha_{kr} e^{j\left(2\pi f_c + \frac{2k\pi}{T_p}\right)\tau_r(i)}$$

where α_{kr} are complex constant and $\hat{n}(i)$ is the noise after match filter. A linear functional relationship of received observation and cardiopulmonary features is appeared by differentiation of $\hat{r}_s(i)$. Because of ultra wideband filtering in match filter, $\hat{n}(i)$ is ultra wideband, and thus this derivative immensely augments the noise such that the extraction of Doppler phase modulation is impossible. $c_{kr}(i)$ are narrowband signals and also linearly extraction of cardio-respiratory factors form their derivative is possible. Hence, in the novel algorithm, narrowband filters are used to estimate $c_{kr}(i)$ and then cardiopulmonary features are extracted from the differentiation of estimated $c_{kr}(i)$. Although, statistical features of $c_{kr}(i)$ are the similar to statistical features of α_{kr} and are stationary in a short time observation, because of time-varying phase, they are statistically non-stationary in long time observation. Narrowband filtering needs long time observation and therefore time-varying filter, that is appropriate for non-stationary signals, must be used in extraction of $c_{kr}(i)$. In the proposed algorithm, a FIR time-varying filter is used. Regarding this filtering model, the estimated $c_{kr}(i)$ can be presented as follows:

$$\hat{c}_{kr}(i) = \sum_{\ell=0}^{M-1} h_{k\ell}^{*}(i)\,\hat{r}(i-k) = \mathbf{h}_k(i)^H \mathbf{r}(i) + n_c(i), \qquad (8)$$
$$\mathbf{r}(i) = \left[\hat{r}_s(i), ..., \hat{r}_s(i-M+1)\right]^T$$

where $\mathbf{h}_k(i)$ is the $M \times 1$ vector of time-varying coefficients of FIR filter; $n_c(i)$ is narrowband noise and superscripts $*$ and H denote conjugate and matrix hermitian, respectively. By determination of FIR coefficients vector, the filtering process is completed. Expanding $c_{kr}(i)$ by optimal orthonormal sequences with maximum energy concentration in a finite sample interval that are known as zeroth discrete prolate spheroidal sequences (DPSS) and considering minimum mean square, the optimum coefficients for FIR filter will be obtained as follows:

$$\mathbf{h}_k^{opt}(i) = \mathbf{F}_k \Phi(i), \qquad (9)$$
$$\Phi(i) = \left[\phi_1(i)\ \phi_2(i)\ ...\ \phi_{\mathcal{M}}(i)\right]^T$$

where $\mathbf{F}_k$ is an $M \times \mathcal{M}$ matrix with entries:

$$f_k(m,\ell) = \phi_\ell(m) e^{j\frac{2\pi k \Delta T}{T_p}m},$$

where $0 \leqslant m \leqslant M-1$; $1 \leqslant \ell \leqslant \mathcal{M}$. $\phi_\ell(i)$ are DPSS orthogonal sequences. $\mathcal{M} = \lfloor 2MB \rfloor$ is the order of expansion, where $B < 0.5$ is the normalized bandwidth of $c_{kr}(i)$ [10]. Details, proofs and error analysis of this estimation and the reason that DPSS are optimum are reported in [10].

The sampling rate ΔT is so fast for the narrowband signal $c_{kr}(i)$, thus these signals are down sampled to sampling rate T_{st}. After decimation, an approximation for differentiation of $c_{kr}(i)$ for $k = 0, ..., K$ are computed as follows:

$$\varepsilon_k(i) = \frac{\left|\hat{c}_{kr}(i) - \hat{c}_{kr}(i-1)\right|}{T_{st}\left(2\pi f_c + \frac{2k\pi}{T_p}\right)} \approx \frac{1}{\left(2\pi f_c + \frac{2k\pi}{T_p}\right)}\left|\frac{dc_{kr}(t)}{dt}\right|_{t=iT_{st}} \qquad (10)$$
$$\approx \beta_b \cos(2\pi f_b T_{st} i) + \beta_h \cos(2\pi f_h T_{st} i) + \hat{n}_c(i),$$

where $\beta_b = |\alpha_{kr}|\,\tau_b f_b$, $\beta_h = |\alpha_{kr}|\,\tau_h f_h$ and $\hat{n}_c(i)$ is the narrowband correlated noise after differentiation. $\varepsilon_k(i)$ is the new signal that expresses the cardio-respiratory features. It is clear that in $\varepsilon_k(i)$, τ_b and τ_h, mentioned in Doppler phase modulation, are intensified by the gain factor f_b and f_h, respectively. Since $f_b \ll f_h$, in $\varepsilon_k(i)$, the portion of heartbeat factor is highly increased and therefore the final estimation of heartbeat rate will be improved. To increase signal to noise ratio, in a next step of processing algorithm, a smoothing procedure is considered as:

$$\bar{\varepsilon}(i) = \frac{1}{K+1}\sum_{k=0}^{K} \varepsilon_k(i) \qquad (11)$$
$$= \bar{\beta}_b \cos(2\pi f_b t) + \bar{\beta}_h \cos(2\pi f_h t) + \bar{n}_c(i),$$

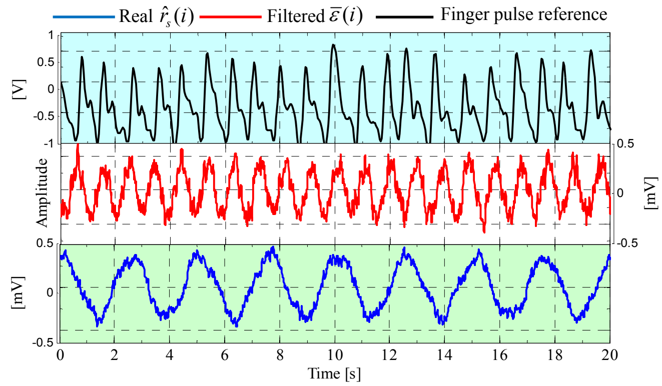
where $\bar{\beta}_b$, $\bar{\beta}_h$ are positive constants and $\bar{n}_c(i)$ is the smoothed noise. In the last step of proposed algorithm, respiratory portion of the achieved signal, $\bar{\varepsilon}(i)$, is filtered out by a tenth-order elliptic IIR band reject filter around estimated breathing rate, $\hat{f}_b$. Finally, the HR is detected in the spectrum of this filtered signal. To find the HRV, time trace of the estimated HR is tracked.

4. EXPERIMENTAL RESULTS

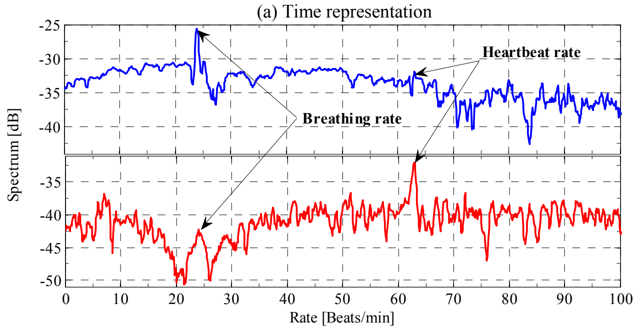
The instrument based UWB radar proposed in [11] with variable power level at the transmitter antenna connector, and two wideband patch antennas provide the suitable observation taken from five healthy adult volunteers in seated position. The processing algorithm is implemented in MATLAB. Parameters, set in the radar setup, is mentioned in table 2. Fig. 2a shows real time representation of sampled received signal as the input of processing algorithm, $\hat{r}_s(i)$, and the output of the algorithm for one of volunteer, compared with a reference signal measuring the pulse from a finger sensor. It can be seen that peaks of output signal are relatively matched to peaks of reference signal, whereas only respiratory effects are completely sensed in the received signal, $\hat{r}_s(i)$. In Fig. 2b, that depicts the spectrum of the input and output signal, the ability of the algorithm in increasing the level of heartbeat factor and removing respiratory factor is obvious. Fig 3a illustrates the time trace of HR measured by reference, novel algorithm and harmonic path algorithm (HAPA), the newly presented signal processing procedure in UWB radar approach. The dominance of our algorithm is clear in this figure. Fig 3b pictures the superiority of the novel algorithm in

Table 2: Parameters of the radar set in the experiment

Parameter	Value	Parameter	Value
center frequency	3.3 GHz	$\mathcal{M}$	70
ΔT	1.5 ns	N	14
T_{st}	300 ms	K	10
Max unambiguous range	12 m	M	512



(a) Time representation



(b) Frequency representation

Fig. 2: a) Time representation of the input and output of the processing algorithm and reference signal. b) The spectrm of the input (upper plot) and output (lower plot) of the processing algorithm.

Table 3: Radar performance in HRV measurement

No.	**Ave. HR [bpm]**		**Relative**	**SDNN [ms]**	
	Radar	**Reference**	**MMSE (%)**	**Radar**	**Reference**
1	62.2	61.5	0.7	21.11	26.13
2	73.6	71.9	1.2	17.48	20.9
3	59.8	60.9	0.9	34.61	30.29
4	79.1	77.7	2.3	43.75	50.01
5	80.2	78.0	2.4	33.39	41.74

an another aspect, the ability of measurement in low quality received signal. In this figure, the relative error in root minimum mean square in different received power level is plotted. HAPA has 4% error in -70 dBm power level, whereas this power for novel algorithm is -80 dBm. The assessment of novel algorithm in measurement of HRV is figured out in table 3. In this table, moreover in the average of HR, the presented algorithm are compared with the reference measurement in the relative error of HRV and in an informative index used in measuring of HRV, standard deviation of normal beat to beat intervals (SDNN) [12], for all five volunteers. The results show that using novel algorithm results in relative error and SDNN with accuracy 2.4% and 8.35 ms, respectively.

5. CONCLUSION

In this work, getting the most out of UWB radar characteristics, a new processing algorithm is presented whose novelty lies in following aspects. First, exploiting both time-varying delay and Doppler phase modulation, the sensitivity to heartbeat motions is improved. Second, by proposing a new time-varying filter and sagely used it to decompose UWB received signal into several narrowband fractions, and using the derivation of these fractions, a new interference free measure is presented. The experimental results show that besides the improvement in performance, the estimation is so accurate that HRV measurement is possible.

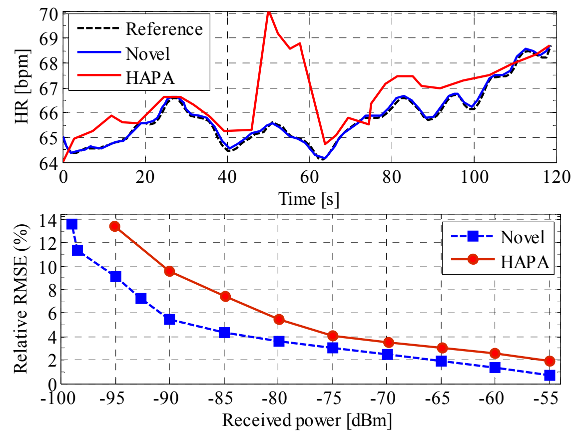


Fig. 3: The upper figure is the time trace of two different approaches comparing with reference signal. The lower figure is the plot of relative error for HR versus received power level.

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