Harmonic Path (HAPA) Algorithm for Non–contact Vital Signs Monitoring with IR–UWB Radar

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***Abstract*—We introduce the Harmonic Path (HAPA) algorithm for estimation of heart rate (HR) and respiration rate (RR) with Impulse Radio Ultrawideband (IR–UWB) radar. A well known result is that a periodic movement, such as the lung wall or heart wall movement, induces a fundamental frequency and its harmonics. IR–UWB enables capture of these spectral components and frequency domain processing enables a low cost implementation. Most existing methods try to identify the fundamental component to estimate the HR and/or RR. However, often the fundamental is distorted or cancelled by interference, such as RR harmonics interference on the HR fundamental, leading to significant error for HR estimation. HAPA is the first reported algorithm to take advantage of the HR harmonics, where there is less interference, to achieve more reliable and robust estimation of the fundamental frequency. Example experimental results for HR estimation demonstrate how our algorithm eliminates errors caused by interference.**

I. Introduction

Vital sign monitoring is fundamental in health care as knowledge of the patient's heart rate (HR) and respiration rate (RR) are essential in identifying clinical disorders and these signs should be monitored consistently and accurately [1]. Currently the most common form of vital sign monitoring in hospitals is pulse-oximetry, in which a sensor is attached to patient's finger tip or probes in case of infants. However for continuous or chronic monitoring, attaching sensors to patient's body causes distress, especially for neonates whose skin is extremely fragile and can be damaged by adhesive probes [2]. Hence, non-contact monitoring is needed not only for patient's comfort but also their safety.

Impulse Radio Ultra-Wideband (IR-UWB) radar is known to be a safe and promising tool for continuous-time, non-contact, non-invasive measurement of HR and RR. Low power IR-UWB is non-ionizing (hence there will be no harm even in continuous monitoring) and has the ability to transmit through obstacles like clothes, bed frame, and blankets [3]. IR-UWB transmitters make negligible interference on other types of radio in its band of 3.1 to 10.6 GHz, and IR-UWB receivers are very robust to interference from other types of radios in this band and multipath [4]. Furthermore, it has extremely high temporal resolution. These advantages make IR-UWB particularly well-suited for home health care, emergency room, intensive care units, hospitals, pediatric monitoring, rescue operation, etc. [3].

The IR-UWB radar system transmits a series of very short and low power electromagnetic (i.e., radio) pulses, typically in the order of nanosecond. When those pulses fall on boundaries between body layers with different dielectric properties such as the skin-air interface, heart wall, and lung wall, the pulses are reflected back to the IR-UWB receiver. The time of arrival of a reflected pulse is related to the round trip distance between the IR-UWB system and the boundary that reflected the pulse. For example, when the radar is placed in the front of the patient, the round-trip distance to the lung wall decreases when the patient inhales and increases when the person exhales. These time-varying times of arrival cause a delay modulation of the IR-UWB received signal, which contains the information about the wall displacements.

II. Related Work

There have been many techniques proposed for non-invasive or non-contact vital signs monitoring. A non-invasive method has been proposed to extract the RR and other related parameters, such as inhale-to-exhale ratio, from the breathing sound [5]. This method requires the sensor to be attached to the chest, causing discomfort and inflexibility. Camera-based monitoring techniques have been proposed for non-contact, non-invasive monitoring of vital signs by detecting the subtle temporal variation of the skin color due to heart beating for HR estimation [6]–[8] and detecting the subtle skin motion due to breathing for RR estimation [6], [7]. However, such methods possess inherent disadvantages. First, they require the subject (or part of the subject) being monitored to be well-illuminated, causing inconvenience, especially during sleeping. Second, they require a line-of-sight between the camera and the part of the subject being monitored. For example a blanket or clothes can easily mask the body's subtle motions due to breathing. Furthermore, this disadvantage prevents the flexibility of sensor locations. Third, privacy is violated.

In contrast, radar-based techniques can offer continuous monitoring, in addition to non-invasiveness and non-contactness. Apparently, Doppler radar and UWB lead the way in radar technologies for vital signs estimation. Doppler radar techniques, however, are known to have the null-point problem [4], hence have limitations if used for biomedical applications. IR-UWB radar does not have this problem and has been demonstrated to be a promising candidate for continuous, non-contact, and non-invasive monitoring of vital signs [3], [4], [9]–[16].

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Spectral analysis is a common approach for radar-based vital signs estimation. Most of the spectral-analysis-based techniques only focus on detecting the respiration and heart fundamental peaks in the spectrum of the radar received signal and overlook their harmonics [16]–[19]. While the strong RR fundamental peak is easy to detect, the HR peak detection is complicated as the strong RR harmonics and (possibly) the intermodulation products between HR and RR may fall into the valid HR frequency range. When these spectral components are near the HR fundamental, they interfere with it because of a well-known phenomenon called "leakage" [20]. Leakage causes slight error in peak locations, i.e., the involved peaks may be shifted away from their true locations, and/or the weak HR peak may get detrimentally attenuated or even completely cancelled. In this case, the peak-selection HR estimation method will have some error which might become very large if the HR is not the highest peak in the valid HR range.

Lazaro [4] proposed a harmonic canceller filter to remove the RR fundamental and its harmonics in the spectrum, and the HR fundamental is estimated to be the global peak over the valid HR range in the "respiration-free" spectrum. However, such method does not work for certain combinations of the subject's true RR and true HR, as given in detail in [9]. A simple example is when the HR is approximately a multiple of the RR, HR fundamental and its harmonics will also be cancelled or attenuated together with the respiration components. In case of RR estimation, body movement might be present, yielding a high amplitude low frequency spectral component that falls into the valid RR range [15], causing leakage to the RR component. If this component is larger than the RR component and there is no other sensor to detect or cancel the movement, the peak-selection technique will mistakenly select this body-movement induced frequency to be the RR estimate.

These problems can be resolved if the useful information in the HR/RR harmonics can be exploited. While the HR fundamental is usually interfered by the RR harmonics, the HR harmonics are relatively free from such interference, as the higher-order RR harmonics are sufficiently weak compared to the nearby HR harmonics. To our knowledge, none of the published spectral-analysis methods have exploited the HR/RR harmonics to improve the estimation of their fundamental frequency. [15] suggested looking for a particular harmonic from which the fundamental frequency will then be computed, but gave no detail on how to locate such harmonic.

The distinct novelty of our study is the development of an algorithm that utilizes not only the fundamental component but also its harmonics to improve the estimation accuracy of the vital signs. We can provide an accurate HR estimate even if the heart fundamental peak is completely missing. Although we will focus on HR estimation throughout our discussion as it is a more challenging task, the algorithm also applies for RR estimation. Our algorithm has been experimentally demonstrated to be robust against the leakage problem and is predicted to be able to provide an estimate of vital signs in the presence of moderate body motion without requiring a body movement canceller. In addition, UWB vital signs estimation methods such as [4], [9], [16], [17], [21] usually require sub-nanosecond sampling intervals, corresponding to sampling rate on the order of tens of GHz, to capture the so-called fast time samples of the received waveforms, in order to recover the respiration and heart beat frequencies which are on the order of only a few Hz (human RR ranges from 0.2 to 0.7 Hz, and HR ranges from 0.75 to 3 Hz). In our method, we first time-gate the signal, to allow radar returns from only the selected span of distances, then we analog-filter the signal to limit the bandwidth to less than 10 Hz. We then sample the signal at a frequency of only 128 Hz, leading to lower cost hardware.

TABLE I
HARMONIC PATH **(HAPA)** ALGORITHM

1) Find all the local maxima (peaks) whose power is above a preselected power threshold.
2) Compute the pair-wise frequency distance between those peaks and retain only the pairs whose pair-wise distance is in the valid human heart rate range.
3) Find the peaks that have approximately equal pair-wise distances and that form a contiguous *path*. A *path* is a set of three or more approximately equi-distant peaks.
4) For each path:
   a) Compute the average inter-peak distance of the path
   b) Perform harmonic test: If the path has at least η nodes whose frequencies are approximately a multiple k of the average inter-peak distance (where "approximately" means the offset is within a specified margin M_k), this path is determined to be a *harmonic path*. The name comes from the fact that the path is formed with the harmonics of some fundamental frequency (the fundamental frequency may or may not be a part of the path). η is an integer greater than or equal to 3 and is user-defined.
5) If only one harmonic path is found, the rate is estimated to be its average inter-peak distance. If more than one harmonic path is found, the rate is estimated to be the average inter-peak distance of the path with highest average power per peak.

The paper is organized as follows. Section III describes the mathematical framework, Section IV explains our rate estimation algorithm based on the inter-distance between the fundamental and its harmonics. Experiment results and conclusion are presented in Section V & VI.

III. SYSTEM MODEL

The mathematical framework and experiment results in [4] show that the spectrum of the radar received signal contains the spectral components centered at multiples of RR, multiples of HR, and their intermodulation products $m \times \text{RR} + l \times \text{HR}$, where m, l are integers. Concrete equations are not reproduced due to length limit, and according to the authors, are not crucial to follow the work presented next.

IV. Harmonic Path (HAPA) Algorithm

The HAPA algorithm was developed based on two simple yet fundamental observations: (1) the HR fundamental and its harmonics, which we will refer to in the sequel as the heart components, are equidistant and are separated by a frequency equal to the HR fundamental; and (2) each heart component is at a multiple of that inter-peak distance. In particular, HAPA will first detect in the received spectrum a *path*, defined as a set of three or more consecutive approximately equally spaced spectral peaks, such that their frequencies are approximately an integer multiple of the average inter-peak distance in terms of frequency; that distance of the most powerful path is the rate estimate. A more detailed description of HAPA algorithm is given in Table I.

HAPA successfully takes advantage of all the significant heart components (the fundamental and the first few harmonics) to improve accuracy of the fundamental frequency estimate. The HR harmonics have less interference from the RR harmonics, therefore they can help average out the peak location error due to leakage and provide an accurate estimate of the HR even when the HR fundamental is completely missing. Another advantage of HAPA is its low processing complexity.

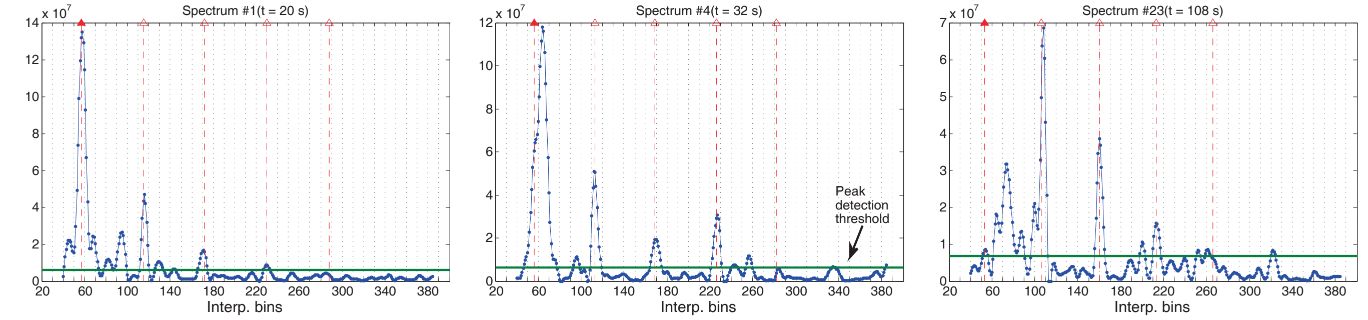
V. Experiment Results

The measurement was conducted with a male subject of 200 lbs and age 42 at rest lying on his back on the top of a mattress. The sensing device used was an IR-UWB radar system developed by Sensiotec Inc. [22], placed under the mattress. The transmit pulses were 13 ns long, centered at 4.2 GHz.

At the receiver of the IR-UWB radar device, the reflected signal was down-converted to baseband and then analog-filtered into the lower respiration band and the higher heart band, using a low pass filter and a high pass filter, respectively. Next, the outputs of each band were sampled at 128 Hz and quantized for subsequent digital signal processing. The recorded data samples, provided by Seniotec Inc., were processed offline for comparison with other estimation methods.

In the digital signal processing step, eight consecutive samples are averaged to reduce noise, producing an effective sampling rate of 16 Hz. We then compute the DFT of each data block containing 256 such 16 Hz samples after DC removal and Hamming windowing. A data block overlaps with its previous block by 75%. The squared-magnitudes of the DFT of two consecutive blocks are averaged, and the average spectrum is interpolated using cubic spline interpolation with ratio four to alleviate the inherent coarseness in spectrum sampling of DFT, providing an approximation of a more finely-sampled version of the true spectrum. Therefore, an interpolated bin n is equivalent to frequency $(60/64)n$ beats/min or bpm.

HAPA is applied to the interpolated, averaged time-dependent spectra, providing real-time estimates of the HR. The preselected threshold in Step 1 of HAPA is chosen to be the 75% percentile of all the interpolated averaged spectrum points. We allow the disparity between the pair-wise (frequency) distances to be within 12 interpolated bins for them to be considered "almost equal" in Step 3. Such disparity margin accounts for the peak location error due to leakage from other spectral components in the spectrum and quantization error. In Step 4, we choose η to be 3, $M_1 = 5$ and $M_k = k + 1$ for $k = 2, 3, \ldots$ and the harmonic test is performed with the rounded average inter-peak distance in unit of interpolated bins. Examples of the interpolated, averaged spectra computed at different time instants are given in Fig. 1. The filled arrow represents the location of the HR fundamental obtained from the pulse-oximeter, the unfilled arrows represent its harmonics, and the horizontal line represents the preselected threshold in our algorithm. In Fig. 1a, the heart fundamental component happens to be the highest peak in the HR range, and both the global peak selection method and HAPA provide accurate estimates. In this case, the heart harmonic path detected by HAPA has frequencies (58, 116, 171, 229), yielding a HR estimate of bin 57 or about 53 bpm. However, in Fig. 1b, the HR fundamental component is not resolved from the nearby strong RR harmonic, and the global peak approach falsely selects this RR harmonic to be the HR estimate. In contrast, HAPA is still able to provide an estimate of the fundamental component. HAPA first detects two harmonic paths whose frequencies are (112, 170, 227) and (112, 227, 335), respectively, and then selects the former since it has highest average power per peak (Step 5). In Fig. 1c,



(a) (b) (c)

Fig. 1. Examples of interpolated averaged spectrum of heart signal. The y-axis is the squared magnitude of DFT values. 1 bpm = $\frac{64}{60}$ interpolated bin. (a) "Clean" HR fundamental and harmonics, (b) Missing HR fundamental, and (c) HR fundamental is severely attenuated.

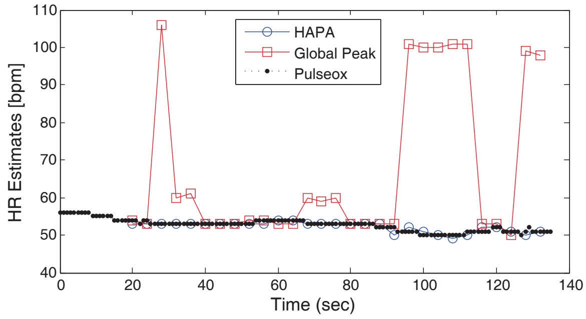


Fig. 2. Heart rate estimates by HAPA and the global peak approach.

the HR fundamental component is severely attenuated due to leakage from other nearby spectral components, including RR harmonics. In this case, the first HR harmonic is the highest peak and therefore selected to be the HR estimate by global peak approach. In contrast, HAPA successfully detects the path (54, 108, 160, 213, 261), formed by the HR fundamental and its harmonic peaks, yielding the correct HR.

The HR estimates given by HAPA and the common global peak approach for the entire 132 seconds of recorded data are shown in Fig. 2. The synchronously obtained pulse-oximeter readings provide ground truth for evaluation of estimation accuracy, and are also shown. As we can see, the HAPA algorithm provides accurate estimates of the HRs whereas the global peak method provides many inaccurate estimates. Specifically the RMSE and RMS normalized error (RMSnE) of HAPA are respectively 0.74 bpm and 1.43% compared to 26.38 bpm or 51.83% of the peak selection approach. The RMSnE is computed by first normalizing each estimation error by the its corresponding HR truth and then applying the root-mean-square operation on these normalized estimation errors.

RR estimation can also benefit from HAPA. In case of an otherwise motionless person, there is negligible spectral leakage from the HR fundamental since the RR fundamental is located in the lower band and is much stronger, thus we do not expect that HAPA will give much different estimation accuracy compared to the common global peak approach. However, regular body motion induces a large amplitude and low frequency component [15] which can cause interference with the RR fundamental. For example, the global peak approach will select body motion component as the RR estimate if it happens to be stronger than the RR fundamental, unless the body motion component is cancelled in advance. In contrast, we predict that HAPA will provide accurate estimate of the RR without the need for body movement cancellation, at least in the moderate body movement case. We plan to investigate this in the future.

VI. Conclusion

We have proposed a novel algorithm, HAPA, for estimation of HR and RR, in which the harmonics are utilized to improve or provide the estimation of the fundamental, which may be otherwise not possible. Experimental results show that our method significantly outperforms the global peak selection method, a common approach that suffers when the heart and respiration components interfere. Specifically, HAPA can provide an accurate HR estimate even when the HR fundamental is missing or has high peak location error due to leakage from nearby RR harmonics. Our algorithm is predicted to improve RR estimation accuracy, especially when the RR fundamental is interfered by the body motion component without requiring a body movement canceller, and this is a topic of our future work. Our algorithm is shown to be practical for a low complexity IR-UWB radar system.

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