Ultra-Wideband Radar Detection of Breathing Rate: A Comparative Evaluation

Nicole Buckingham[a] and Denis Gračanin[b]
Department of Computer Science, Virginia Tech, 2202 Kraft Drive, Blacksburg, VA 24060, U.S.A.

Keywords: Biosensors, Breathing Rate, Human Computer Interaction, Remote Sensing.

Abstract: The cost of a medical grade breathing rate monitors can be prohibitive. However, commodity ultra-wideband (UWB) radar based device can be used to detect breathing rate for health monitoring applications in homes. We identified several research challenges, including high cost and functional limitations based on the user's location, orientation, and movement, as well as dependency on system placement and vulnerabilities in signal processing methods. We performed a comparative evaluation for a commodity UWB radar based device, Walabot, to determine its feasibility for health monitoring applications. The data was processed using two breathing rate derivation techniques: Fast Fourier Transformation (FFT) and Peak Detection. The results support feasibility of Walabot as a commodity breathing rate monitor for health monitoring in homes.

1 INTRODUCTION

Human body produces a variety of physiological signals. Four important signals are the heart rate, breathing rate, temperature and blood pressure. These signals are known as the vital signs because they provide a well rounded indication of the overall state of the body. Health monitoring focuses on measuring various physiological signals and processing these signals to determine health status.

Heart rate is controlled by the rate at which the sinoatrial node creates electrical impulses. Resting heart rate depends on age, gender and exercise level, but an increase in resting heart rate can reveal declining heart health and increase in risk of heart attack.

The respiratory system is responsible for the breathing mechanism. The rate at which breathing occurs is specifically controlled by the respiratory pacemaker. Variation in breathing rate are commonly associated with conditions like asthma, anxiety, pneumonia, lung disease and congestive heart failure (Cleveland Clinic, 2019).

Traditionally, heart rate is used to determine risk of heart attack and failure. However, studies have shown that breathing rate is a more accurate way to detect or predict heart conditions, such as cardiac arrest (Cretikos et al., 2008).

[a] https://orcid.org/0000-0003-2558-5861
[b] https://orcid.org/0000-0001-6831-2818

Contactless health monitoring systems use sensor(s) placed in the vicinity of the user but without any direct user contact. The goal is to provide the same capabilities of wearable technology, but with the added benefits of increased comfort. This is significant for health applications because the users may not always be able to correctly use wearable devices.

The effectiveness of health monitoring systems is affected by the environment in which they are used. Contactless devices, in particular, can benefit from environments such as Smart Built Environments (SBE) that provide data collection and analysis services integrated within a control infrastructure (Tasooji et al., 2018). The cost of a medical grade breathing rate monitors can be prohibitive for use in SBEs (e.g., smart homes). However, there are technologies that can be used for contactless breathing rate monitors.

Ultra-wideband (UWB) devices use a wide spectrum of low energy radio frequency signals to determine the chest displacement caused by breathing. The measurements can be very accurate but also can be sensitive to noise and limited by user location. UWB devices have a potential to enable ubiquitous breathing rate monitoring by continuously monitoring users and predicting health events in real time. UWB based system have limitations in terms of the coverage area and the user's position and orientation. However, with the availability of commodity UWB devices such as Walabot (Walabot, 2020), it is important to understand those limitations. We

Buckingham, N. and Gračanin, D.
Ultra-Wideband Radar Detection of Breathing Rate: A Comparative Evaluation.
DOI: 10.5220/0010346906770684
In *Proceedings of the 14th International Joint Conference on Biomedical Engineering Systems and Technologies (BIOSTEC 2021) - Volume 5: HEALTHINF*, pages 677-684
ISBN: 978-989-758-490-9

have conducted a comparative evaluation of Walabot and a medical grade contact breathing rate monitor (MindWare Mobile Impedance Cardiograph (MindWare Technologies LTD, 2013)) using 4 procedures: breathing rate variation, horizontal placement variation, vertical placement variation and user movement variation. The results support feasibility of Walabot as a commodity breathing rate monitor for health monitoring in homes.

2 RELATED WORK

Breathing, or respiration, is a vital mechanism that all humans perform subconsciously in which air circulates in and out of the lungs, in order to provide oxygen to the body. The rate and volume of air moved is controlled by the respiratory pacemaker, which is located in the medulla of the brainstem.

Tidal volume is the volume of air intake during inspiration. Typical tidal volume is dependent on the individual but a volume of 400–500 mL is considered healthy (Hallett and Ashurst., 2019). In healthy adults, tidal volume is quite consistent between breaths, but infrequent changes in air intake due to yawns or signs are considered normal. Tidal volume may also change in situations where breathing rate changes, such as exercise (Braun, 1990).

Respiration devices typically measure breathing rate and tidal volume. Breathing (respiration) rate is number of breaths per minute (bpm). Typical resting breathing rate varies per person but a rate of 12-20 bpm for an adult is considered normal, whereas a resting rate over 25 or under 12 is considered unhealthy. Various health conditions affect resting breathing rate, such as asthma, anxiety, pneumonia, congestive heart failure and lung disease (Cleveland Clinic, 2019). Changes in breathing rate in hospital patients is correlated with an increase in mortality rate.

One limitation of breathing rate is that the measurement can be affected by the time over which is it measured. Some studies extract breathing rate from a shorter time, such as 15 seconds. However, this measurement is subject to inaccuracy because breathing rate can change between measurement periods due to changes in air intake, as described in the last paragraph. Studies have shown that the time period with the least variability is one minute (Evans et al., 2001).

There are a variety of wearable medical devices on the market and even more being developed and tested currently. These devices appear as normal clothing or accessory articles, but they consist of small sensors that can read a variety of biometric data.

From a physiological perspective, breathing rate offers a unique opportunity for wearable devices. In contrast to heart rate, which is measured using internal signals, physical signals caused by breathing rate can be recorded using audio, airflow or the low frequency mechanical signal caused by small chest inflections (Furtak et al., 2013). These physical effects have made breathing rate a popular subject of wearable device research.

Contactless monitoring of breathing rate is a novel topic of research. Typical contactless methods use thermal sensors, acoustic changes, lasers or radar. Radar methods encompass techniques that utilize microwave or radio frequency (RF). Such methods rely on the same idea: a stationary person's breathing rate matches the phase shifts of signals reflected off the person (Lin, 1975). These methods are typically categorized as continuous-wave (CW), frequency-modulated continuous-wave (FMCW), impulse radar, and ultra-wideband (UWB) (Brüser et al., 2015).

3 METHODS

UWB radar emits a wide spectrum of low energy RF that can be used to capture the low frequency mechanical signal caused by small chest inflections during the breathing cycle. Our goal is to investigate the following question: "*Can UWB radar be utilized to enable a fully functional breathing rate monitoring system for an active user in a confined SBE?*"

For this work, we define a fully functional system as one that derives accurate breathing rate data within 10% of the true breathing rate. Further, such a system should maintain accuracy at all times in which the user is present in the confined space and should not be limited by user location, orientation or movement. The specific area covered by the system is dependent on the particular system but the minimum area should cover a standard bedroom or living room. For the scope of this work, the system will be restricted to one user within range at a time.

The ideal system would collect breathing rate for each user, detect distress events, alert the user's connected devices and even call for emergency help when necessary. However, we focus on the breathing rate detection system only. These potential functionalities could be evaluated in future work on devices that are deemed fully functional.

The reviewed UWB devices show promising results but are not considered fully functional because they do not meet the requirements described above. Specifically, these systems are vulnerable to noise or limited by user. There are several important challenges to overcome. The details and severity of each

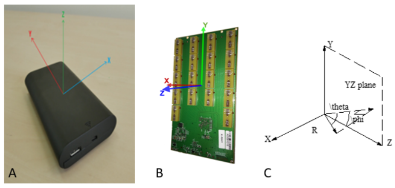

Figure 1: Walabot coordinate systems: **A)**Walabot device and its Cartesian coordinate system. **B)**Walabot antenna array. **C)**Cartesian and spherical coordinate systems.

challenge is specific to the device and system design, but the challenges for the current systems are:

1. **Cost:** They are expensive, thus not affordable for an average household.
2. **System Placement:** Dependance on the placement of the transmitter and receiver device(s).
3. **Signal Processing:** Deriving breathing rate from the received signal is sensitive to noise.
4. **User Location:** The user has to be in particular locations relative to transmitter / receiver device(s).
5. **User Orientation:** The user has to be oriented in particular angles relative to the transmitter and receiver device(s).
6. **User Movement:** Increased sensitivity to noise or inaccuracy when the user is moving.

3.1 Design

We conducted a comparative evaluation of Walabot (Figure 1) and a medical grade contact breathing rate monitor (MindWare Mobile Impedance Cardiograph (MindWare Technologies LTD, 2013)) to determine feasibility of UWB devices for home health monitoring systems. There are 6 design criteria.

1. Only one user can be in the designated space.
2. The user must use different breathing rates to test the full range of possible breathing rates.
3. There must be at least one trial where the user must remain still and there must be at least one trial where the user must move.
4. The confined space should be approximately the size of a standard living room or bedroom.
5. Walabot must be placed in a variety of locations and orientations with respect to the user's chest.
6. There must be at least one trial where the user sits, stands and walks in place.

The main goal was to analyze the effectiveness and limitations of a commodity UWB radar based device as a breathing rate monitoring system against a medical grade wearable breathing rate monitor. The secondary goal was to determine how well Walabot meets the identified requirements. The steps are:

1. Analyze and discuss of the methodology used to gather breathing rate provided by Walabot API.
2. Develop a breathing data acquisition for Walabot.
3. Develop a signal post processing script to compute the breathing rate of data simultaneously generated by Walabot and Mobile devices
4. Design a comparative evaluation to test the accuracy and limitations of Walabot.
5. Provide the comparative evaluation results and discuss the accuracy and limitations of Walabot.
6. Discuss Walabot in the context of the posed research question and challenges.

The research question and daily living emphasis informed the first four design criteria. **First:** only one user was involved in the data acquisition session due to COVID-19 situation. **Second:** the user was told to use the following breathing rates during different sets of data acquisition: normal, deep and fast. To ensure the deep and fast rates were within reason, the user aimed to reach 8–12 bpm during deep breathing and 15–20 bpm during fast breathing. **Third:** the participant was instructed to remain still for a set time and to perform a predetermined movement for a set time during the data acquisition session. The chosen positions and movement were specified in a later design decision. **Fourth:** the participant and the device were placed in a specified area, within the size of a standard living room or bedroom. This stems from the assumption that the device would be placed in a common living area to be utilized frequently. A sketch of the layout of the room can be seen in Figure 2 top and a picture of the actual room is illustrated in Figure 2 bottom. The user was positioned 60 cm away from Walabot along Z axis and the exact positioning of Walabot was varied within this layout throughout data acquisition.

4 EVALUATION RESULTS

The comparative evaluation was carefully designed in such a way that allows for insight into both the research question and the identified criteria. Additionally, it was ensured that the testing setup should reflect how the device would be theoretically used in a real living space in order to make the data useful for real world applications.

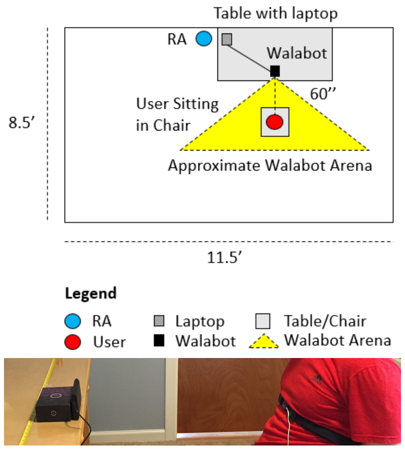

Figure 2: **Top:**A sketch of the testing setup. **Bottom:**A picture of the testing setup.

4.1 Raw Data Comparison

Walabot and Mobile data were aligned relative to set number, time and frequency. However, there are two important differences in Walabot and Mobile: magnitude and stability. The magnitude of the two signals shouldn’t be directly compared. Mobile raw signal does not have a significant DC component, whereas Walabot has a large DC component (Figure 3 top). Once the DC components are removed, the difference in magnitude in the signals was clearly noticeable. To eliminate the magnitude differences, the signals were normalized signals between -1 and 1 for the best visual illustration of the signals. The shape should resemble the expected respiration sine wave form with peaks and valleys according to inhaling and exhaling respectively and the frequency should correspond to the breathing rate.

With regards to stability, Walabot data is noticeably less stable than Mobile data. Kilani fixed this issue by averaging the collected energy values over a sliding window of 5 samples (Kilani, 2017). Instead, this signal can be stabilized by smoothing. using a sliding window. The effects of normalization and the smoothing are shown in Figure 3 middle.

The simplest way to compare the data is to visually compare the shape of the graphs for each device. In an effort to quantify this comparison, we determined the a correlation between the two signals. However, the correlation coefficient is not a determination of signal accuracy because the concept of accuracy in this work is based on the breathing rate calculated from the signals. Further, the signals may have a low correlation coefficient but both be highly accurate. Instead, the number of extrema within each signal is a good indication of what the breathing rate calculation will be.

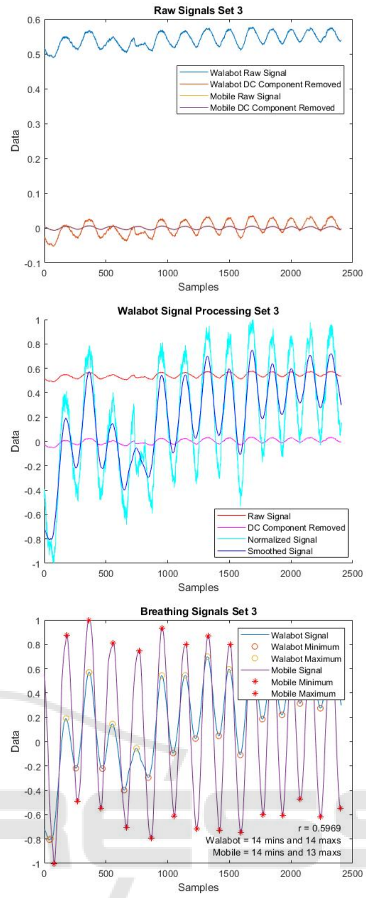

Figure 3: **Top:**An example of raw Walabot and Mobile signals. **Middle:**An example of the effects of normalization and smoothing of Walabot signal. **Bottom:**An example of processed Walabot and Mobile signals and their extrema and their correlation coefficient.

An example of this signal comparison is shown in Figure 3 bottom. The correlation coefficient is shown in the bottom right hand corner of the graph. In this example, the coefficient is 0.68. Visually, the signals appear to have very similar frequencies and very similar number of minimums and maximums between Walabot and Mobile signals. The correlation coefficient is not 1 since the shapes of the two signals vary.

4.2 FFT Signal Analysis Technique

Extracting breathing rate from raw data is a complex process that is still being researched today. In fact, there are over 100 methods for extracting breathing rate from ECG and PPG signals alone (Charlton et al., 2016). Signal processing of radar data to find breathing rate is a popular topic is research currently (Li et al., 2016; Taheri and Sant’Anna, 2014). There are many proposed extraction methods, but there is no gold standard technique for Walabot at this point.

As a starting point, we attempted to replicate the calculation method used by Mobile. Specifically, Mobile uses a low pass filter, then calculates a respiration trend once per second using the FFT technique over a sliding window of 5 seconds.

Next, the sliding window was defined. Initially, the window was set as 5 times the fs value, or 5 seconds, and also tested at 7 and 10 seconds.

During each iteration of the loop, a FFT was performed on that window of the filtered data. Commonly, this is done in a sliding window to see smaller time periods and to evaluate the changes over time (Gunasekara, 2017).

The DFT was used to find the fundamental breathing frequency. The data contains very few data points which causes the resolution of the DFT to be very low and the calculated breathing rate is not very accurate. Spline interpolation was used to increase the resolution. The points before, on and after the max index were cubically interpolated.

When the sliding window finishes, an extra filtering step is performed to clean up the breathing rate calculations. First, the first and last bpm calculations are deleted. Then, any calculations below 6 or above 25 are deleted to disregard any values deemed unrealistic. These numbers were derived from the estimation that the average breathing rate for a health adult is between 12 and 20 bpm (Cleveland Clinic, 2019).

Then, the average breathing rate was calculated throughout the set. The accuracy of the average breathing rate is $100*(1-\frac{|\text{Calculated bpm}-\text{User Reported bpm}|}{\text{User Reported bpm}})$.

To illustrate the results, the calculated bpm and the averages are shown (Figure 4 top). The average breathing rate was 12.84 bpm for Walabot and 12.53 bpm for Mobile. The trends reported by Mobile for the breathing rate is shown (Figure 4 middle). The average reported breathing rate was 11.21 bpm.

The calculated average breathing rate for Mobile data was 1.33 bpm above the reported average breathing rate. The user counted roughly 13.5 bpm. The reported breathing rate was not within 10%, while the calculated breathing rates for Walabot and Mobile were both within 10% (Figure 4 middle).

4.3 Peak Detection

We implemented a second breathing rate extraction method. Yang et al. proposed the use of time-domain peak detection in order to extract breathing rate within a window of one breathing cycle, or 5 seconds. This technique was tested with Doppler radar, an ECG and a respiration band and yielded highly correlated results between the two contact sensors (Yang et al., 2017). This test is quite similar to our work as it used both radar and a respiration band. Additionally, this technique was deemed a viable option based on visual observation of our collected data.

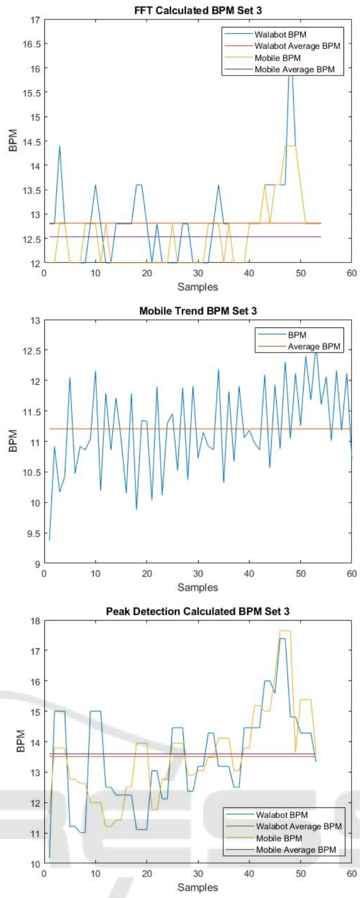

Figure 4: **Top:** An example of bpm calculated from Walabot and Mobile raw data using the FFT technique. **Middle:** An example of bpm provided by Mobile. **Bottom:** An example of bpm calculated from Walabot and Mobile data using the Peak Detection Technique.

We first implemented the same lowpass filter used in the FFT technique. Yang et al. suggested the use of a bandpass filter, but we used a lowpass filter to maintain consistency between the post processing and Mobile processing suggestions (Yang et al., 2017).

Next, the sliding window from the FFT technique was implemented again. To fully test and compare our processing techniques, windows with 5, 7 and 10 seconds were tested. For each window we located the indices with the local minimum and maximum points. The period was then calculated by doubling the time between the minimum and maximum points. Finally, the breathing rate was calculated as $60*\frac{fs}{period}$.

When the sliding window finishes, the extra filtering step that was implemented in the FFT technique

Table 1: Mobile reported bpm results.

Set	User Reported bpm	Mobile Reported bpm	Mobile Reported bpm Accuracy
1	13	11.1502	85.77%
2	15	11.6261	77.51%
3	13.5	11.2050	83.00%

is used again here to clean up the breathing rate calculations. Then the average breathing rate and corresponding accuracy's are computed. This procedure explained was used to calculate the breathing rate for both Walabot and Mobile signals for each set. To illustrate the results, the calculated bpms and the averages are shown (Figure 4 bottom). In this set, the average breathing rate for Walabot was found to be 13.38 bpm and the average breathing rate for Mobile was 13.68 bpm. With this extraction technique, the calculated breathing rate for Walabot and Mobile were both correct within 10% of the user reported 13.5 breaths. Further, this technique yielded breathing rate calculations with significantly higher accuracy than the FFT calculations. Additionally, the trends reported for the breathing rate of Mobile data are shown (Figure 4 middle). The average reported bpm is 11.21 bpm for this data set which is not within 10% of the user counted breaths of 13.5 bpm.

4.4 Accuracy Comparison

Three sets of data were taken on a 23 year old male. During each set, the user was asked to breath normally. The user reported approximately 13, 15 and 13.5 bpm for sets 1, 2 and 3 respectively.

Another accuracy metric uses Mobile to Walabot bpm ratio as percentage to compare the similarity of the two calculations. When this value is within 10% of 100%, the two calculations are considered statistically equivalent and this indicates high correlation of signals and high stability of the associated breathing rate derivation technique.

Table 1 shows the average Mobile reported breathing rate and the respective accuracy's. Prior to testing, Mobile measured breathing rate was expected to be the most accurate. However, the accuracy's are all significantly lower than 90%. Further, none of the reported averages are within 10% of the expected value. Due to this testing, Mobile reported breathing rate values were not used in further testing.

Table 2 shows the FFT calculations using window sizes of 5, 7 and 10. Calculations with a window size of 5 seconds are within 10% of the expected value. The calculations for Mobile data with a window size of 7 and 10 seconds are within 10% of the expected value. However, one calculation for Walabot with a window size of 7 seconds and two calculations with a window size of 10 seconds are not within 10% of the expected value. The second set is only 55.24% accurate when using a window size of 10 seconds. This is an interesting observation given that increasing the window size increases the resolution, which was thought to improve the accuracy. The ratios are within 10% for a window size of 5 seconds, but the ratios are outside of the 10% bounds for larger windows. Due to these results, a window size of 5 seconds was used in further testing of the FFT technique. Table 3 shows the Peak Detection technique results using window sizes of 5, 7 and 10. All results are within 10% of the expected value and the ratios are all within 10%n. An interesting observation is that the window size does not have a significant impact on the accuracy of the calculation. In order to maintain consistency with the FFT technique, a window size of 5 seconds was used in further testing.

5 DISCUSSION

The comparative evaluation is a necessary first step to understand the accuracy and limitations of Walabot breathing rate measurements. Six criteria were used: cost, system placement, signal processing, user location, user orientation, and user movement.

Cost: The least expensive version of Walabot is less than $100 and is considered affordable for an average household as a health monitoring device.

System Placement: Walabot has an advantage because the antennas are all located within the device (rather than having a separate transmitter and receiver), making it easy to deploy.

Signal Processing: Two breathing rate derivation techniques were used, FFT and Peak Detection. The results were compared against the user reported breathing rate (13, 15 and 13.5 bpm) to obtain an accuracy measurement. A window size of 5 seconds was chosen for both techniques because then all the results were within 10%. During the breathing rate variation procedure of the comparative evaluation, an issue with the window size was uncovered. Essentially, using a window size of 5 seconds is ideal for breathing rates between 12 and 20 bpm, but this window size is too small for breathing rates lower than 12. This is caused by fact that the calculation techniques require a window size of at least one breath to accurately determine breathing rate, but less than one breath is taken during a 5 second window when the breathing rate is lower than 12 bpm.

When the window size was 10 seconds, the accu-

Table 2: FFT technique results.

Set	Window Size	User Reported bpm	Mobile bpm Calculation	Mobile bpm Accuracy	Walabot bpm Calculation	Walabot bpm Accuracy	Mobile to Walabot Ratio
1	5	13	12.37	95.16%	13.35	97.32%	92.66%
	7	13	13.20	98.48%	11.70	90.03%	112.82%
	10	13	12.06	92.75%	9.36	71.96%	128.85%
2	5	15	13.69	91.27%	13.99	93.28%	97.86%
	7	15	16.49	90.04%	11.41	76.04%	144.52%
	10	15	14.87	99.10%	8.29	55.24%	179.37%
3	5	13.5	12.53	92.84%	12.81	94.92%	97.81%
	7	13.5	13.87	97.27%	13.74	98.25%	100.95%
	10	13.5	12.85	95.18%	12.16	90.04%	105.67%

Table 3: Peak detection technique results.

Set	Window Size	User Reported bpm	Mobile bpm Calculation	Mobile bpm Accuracy	Walabot bpm Calculation	Walabot bpm Accuracy	Mobile to Walabot Ratio
1	5	13	12.97	99.74%	13.26	97.97%	97.81%
	7	13	12.99	99.91%	13.12	99.09%	99.01%
	10	13	13.07	99.45%	13.27	97.93%	98.49%
2	5	15	15.05	99.68%	15.05	98.21%	100%
	7	15	15.15	99.02%	14.89	99.25%	101.75%
	10	15	15.25	98.33%	14.66	97.75%	104.02%
3	5	13.5	13.60	99.27%	13.51	99.94%	106.66%
	7	13.5	13.59	99.36%	13.52	99.86%	100.52%
	10	13.5	13.60	99.30%	13.51	99.96%	106.66%

racy for breathing rate lower than 12 bpm increased. However, increasing the window size decreases the accuracy of calculations during sets with breathing rates above 12 bpm. The original window size of 5 seconds was used for the rest of testing during the comparative evaluation. Adding a window size adjustment capability could improve the accuracy of the calculations for any breathing rate.

User Location: Walabot is expected to be functional whenever a user is within the Arena specified within the data acquisition software by the values of R, ϕ and θ. This allowed for the simplification of the location challenge because the R range could simply be set based on the user's locations during the study. Specifically, the range of R in the Arena was set as 20 to 80 cm because the user was always positioned 60 cm away from the device in Z direction. However, the minimum value of R is 1 cm and the maximum value is 1000 cm, which allows Z axis to be set such that it covers a typical living room or bedroom.

User Orientation: The horizontal and vertical placement variation procedures were followed. Within each of these procedures, Walabot was angled -45, -22.5, 0, 22.5 and 45 degrees away from the user's chest in the X or Y axis. The results showed relatively high accuracy of breathing rate computed by the FFT technique. The results for all locations were all above 85% for both horizontal and vertical testing when the reported breathing rate was above 12 bpm. The Peak Detection technique had quite low accuracy. This suggest that the FFT technique can provide a more accurate breathing rate when the user is not located directly in front of Walabot. However, these results were not all within 10% accuracy. Consequently, Walabot coupled with the signal processing techniques developed do not overcome the user orientation challenge at this point. It is important to note that the θ and ϕ values determine the cone shape of the Arena in X and Y axes. For this work, the ranges for both θ and ϕ were set as -1 to 1 because this allowed for the highest sampling frequency. Further testing should be done with higher θ and ϕ values to determine if the user orientation challenge is minimized by a larger Arena size.

User Movement: The user stood against a wall, in an open area and walked in place. Walabot signal was very noisy during moving trials, which caused the Peak Detection technique accuracy to drop below 90%. The FFT technique was able to accurately compute breathing rate regardless of the user's movement during the trials when the reported breathing rate was above 12 bpm. This suggests that Walabot with the FFT techniques overcomes the user movement challenge. Results from the horizontal, vertical and user movement variation procedures revealed the high accuracy and reliability of the FFT technique when the reported breathing rate is above 12 bpm.

FFT technique yielded results with higher accuracy than the Peak Detection technique. The primary reason for this is that the FFT method is not significantly affected by the noise in the shape signal, while the Peak Detection method is highly affected. For this reason, the FFT method should be the primary focus in future testing. However, the FFT technique is not sufficiently robust at this point. An adjustable window size can increase the accuracy of the FFT technique.

6 CONCLUSIONS

The development of a robust and fully functional UWB radar based system has the potential to provide accurate monitoring of breathing rate. However, current UWB radar based systems have issues which hinder their accuracy or reliability. Six criteria were identified: cost, user location, user orientation, user movement, system placement and signal processing. We designed and performed a comparative evaluation in which data was collected by following four procedures: breathing rate variation, horizontal placement variation, vertical placement variation and user movement variation. Results from this study were promising and suggested a high potential for Walabot coupled with the FFT technique. Specifically, it was determined that this system meets the cost, user location, and system placement criteria. However, further testing is required to determine if the system can fully meet the user orientation, user movement and signal processing criteria. The results support feasibility of Walabot as a commodity breathing rate monitor for health monitoring in homes.

REFERENCES

Braun, S. R. (1990). Respiratory rate and pattern. In Walker, H. K., Hall, W. D., and Hurst, J. W., editors, *Clinical Methods: The History, Physical, and Laboratory Examinations*, chapter 43, pages 226–230. Butterworths, Boston, 3 edition.

Brüser, C., Antink, C. H., Wartzek, T., Walter, M., and Leonhardt, S. (2015). Ambient and unobtrusive cardiorespiratory monitoring techniques. *IEEE Reviews in Biomedical Engineering*, 8:30–43.

Charlton, P. H., Bonnici, T., Tarassenko, L., Clifton, D. A., Beale, R., and Watkinson, P. J. (2016). An assessment of algorithms to estimate respiratory rate from the electrocardiogram and photoplethysmogram. *Physiological Measurement*, 37(4):610–626.

Cleveland Clinic (2019). Vital signs. https://my.clevelandclinic.org/health/articles/10881-vital-signs. [last accessed 17 December 2020].

Cretikos, M. A., Bellomo, R., Hillman, K., Chen, J., Finfer, S., and Flabouris, A. (2008). Respiratory rate: the neglected vital sign. *The Medical Journal of Australia*, 188:657–659.

Evans, D., Hodgkinson, B., and Berry, J. (2001). Vital signs in hospital patients: a systematic review. *International Journal of Nursing Studies*, 38(6):643–650.

Furtak, N. T., Skrzetuska, E., and Krucinska, I. (2013). Development of screen-printed breathing rate sensors. *Fibres and Textiles in Eastern Europe*, 21(6):84–88.

Gunasekara, A. (2017). Contactless estimation of breathing rate using UWB radar. Master's thesis, School of Electrical Engineering and Computer Science University of Ottawa. [last accessed 17 December 2020].

Hallett, S. and Ashurst., J. V. (2019). *Physiology, Tidal Volume*. StatPearls Publishing, Treasure Island, FL.

Kilani, R. (2017). Testing the viability of Walabot for breathing detection. https://www.hackster.io/Ruleof2/testing-the-viability-of-walabot-for-breathing-detection-b0a14f. [Last accessed 17 December 2020].

Li, W., Tan, B., and Piechocki, R. J. (2016). Non-contact breathing detection using passive radar. In *IEEE International Conference on Communications*, pages 1–6, Kuala Lumpur, Malaysia.

Lin, J. C. (1975). Noninvasive microwave measurement of respiration. *Proceedings of the IEEE*, 63(10):1530-1530

MindWare Technologies LTD (2013). Mindware mobile impedance cardiograph product manual. http://downloads.mindwaretech.com/Hardware/MindWare Mobile Impedance Cardiograph 1.0.0-1 Manual.pdf. [Last accessed 17 December 2020].

Taheri, T. and Sant'Anna, A. (2014). Non-invasive breathing rate detection using a very low power ultra-wide-band radar. In *IEEE International Conference on Bioinformatics and Biomedicine*, pages 78–83.

Tasooji, R., Dasgupta, A., Gračanin, D., LaGro, M., and Matković, K. (2018). A multi-purpose IoT framework for smart built environments. In Rabe, M., Juan, A. A., Mustafee, N., Skoogh, A., Jain, S., and Johansson, B., editors, *Proceedings of the 2018 Winter Simulation Conference*, pages 4240–4241.

Walabot (2020). Walabot DIY. https://walabot.com/diy. [Last accessed 17 December 2020].

Yang, X., Sun, G., and Ishibashi, K. (2017). Non-contact acquisition of respiration and heart rates using doppler radar with time domain peak-detection algorithm. In *2017 39th Annual International Conference of the IEEE Engineering in Medicine and Biology Society (EMBC)*, pages 2847–2850.