Motion Artifact Reduction for Respiratory Monitoring: a Multichannel Ultrasound Sensor for Diaphragm Tracking

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Abstract—**The ability to assess the respiratory rate reliably is vital in the application of remote health monitoring. A newly developed pulmonary monitoring system based on diaphragm wall motion tracking is evaluated under human physical activity. Diaphragm excursions are tracked by a designed ultrasound sensory system utilizing three ultrasound PZT5 piezo transducers. We assess the accuracy and reliability of this system in monitoring the diaphragmatic function and its contribution to the respiratory workload. We also examine inertial and pulse-oximetry sensors as two alternative methods. Measurements are compared to a spirometer as the gold standard. The tests are designed in this study to investigate the performance of employed sensors in both the stationary and non-stationary human body situations. We conclude that by the direct tracking of the diaphragm motion, the proposed ultrasound system is fundamentally robust to motion artifacts. Outstanding results were obtained for the proposed system from six subjects in six different test models with an average sensitivity, specificity and false alarm of 89.7%, 93.1% and 6.3%, respectively.**

Index Terms—**Respiratory monitoring; Diaphragmatic excursion; Ultrasound; Motion artifact; Pulse oximetry sensor; Respiratory sinus arrhythmia (RSA).**

I. Introduction

Respiratory rate (RR), or the number of breaths per minute, is an important clinical sign representing ventilation. A change in RR is often the first sign of deterioration in human vital signs as the body tries to balance oxygen delivery to the tissues. In a research by Cretikos *et al* [1], over half of examined patients suffering a serious adverse event on the general wards (such as a cardiac arrest or ICU admission) had a respiratory rate of 24 breaths/minute and more. These patients could have been identified as high risk up to 24 hours earlier. In another research by Jonsson *et al* [2], they found that early detection and monitoring of changes in vital signs, particularly RR, could help to anticipate and identify respiratory failure, which is the most common cause of admission to hospital's intensive care unit (ICU). Monitoring ventilation during sleep, in terms of hypopneas and apnoeas (i.e. finding the occurrence of periods with low or no ventilation), is a necessity in the sleep disorder screening. Therefore, there is an essential need to have a reliable respiratory monitoring system [3].

Wearable technologies such as wrist-worn reflectance PPG sensors, are used to monitor physiological parameters or vital signs such as heart rate, blood oxygen level and respiratory rate. However, not all wearable technologies have been successfully utilized in the health-care practices. Although wearable technology has the ability to recognize patients at risk, many devices do not acquire the signals in a sufficiently reliable way for clinical use. There is a lack of robustness in the estimate of vital signs that wearable sensors produce. One reason could be the significant motion artifacts affecting the quality of recorded signals which can decrease the accuracy of vital sign estimations.

Prolonged breath monitoring during different body motions and positions is important in many applications for personal health tracking or in hospitals. Numerous non-invasive devices for respiration monitoring have been proposed. These monitoring devices could be divided into direct and indirect methods. Direct methods obtain the breathing operation information directly from the pulmonary system. Examples are flow meters based on pressure transducer [4], thermal [5] and ultrasound flow meters [6]. However, these methods are not eligible as a comfortable gadget for long-term respiratory monitoring as the user has to breath into a pipe or a sensor has to be placed near the nose or mouth. Another example of a direct method is the chest motion analysis using inertial sensors [7], [8]. Since this method relies on the chest or abdomen motions caused by breathing, it is susceptible to false readings due to human movements.

Photoplethysmography (PPG) sensor is the most popular and inexpensive method for heart rate monitoring [9] and in some applications it is used for respiratory monitoring [10] [11] [12]. This indirect method is based on small blood pressure variations that the pulmonary system applies [13]. The signal baseline value of this sensor is extremely sensitive to motions, which results in variations to the respiratory and heart signals. This

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impact is dominant over the respiratory signal [14] [15] [16]. However, some researchers did not consider the portion of the signal contaminated by motion artifacts [10] [11]. In addition, it is not favorable to carry a sensor on the finger-tip, earlobe or forehead [17] [18]. Another well-known method for respiratory monitoring, called electrical impedance tomography (EIT), obtains the respiratory signal through impedance changes in thorax [19] [20]. This method requires the connection of at least four conductive electrodes to the skin. Moreover, the movement-induced artifacts on its data cannot be compensated by filtering methods [21] [22].

Ultrasound technology has been utilized for imaging the human body for over half a century. Functioning of the diaphragm using ultrasound imaging machine was assessed by Zambon *et al.* [23], but it required a technician to hold the probe. Ultrasound technology (US) is inexpensive, safe, real-time and can be used as a direct method for pulmonary monitoring. Ultrasound imaging systems with an array of piezo transducers are widely used in diagnosis and imaging [24]. In this study, we utilized only one sensor, consisting of three PZT5 piezo disk transducers as a 1-D tracker of internal organs. The internal organ is the diaphragm muscle observed from the zone of apposition. The transducers operate in the thickness mode with the resonant frequency of 2.2 MHz. A designed multichannel mixed-signal system generates and records ultrasound pulses and its echoes as a function of intensity and time (depth), called ultrasound B-mode operation. Intensity and time of flight (ToF) of reflected echoes from observing organs reveal the position of the organ in one dimension. To determine the velocity and motion of the diaphragm, the system averages the amplitude of reflections over the time in the specified depth. Previously, an ultrasound based system based on one transducer proposed by the authors [16]. The system showed a good correlation in respiratory detection versus the gold standard in different breathing patterns. However, the system witnessed a reading error due to the sensor displacement in situations where the subject moved or stretched his body a lot. In this paper, we propose an improved sensor versus the previous one and evaluated it under different body kinematics. The system shows less sensitivity to body motion than the previously proposed system. In Section II, we explain where the sensor should be placed on the body and how the system is going to be evaluated with other methods. Section III describes the proposed multichannel ultrasound system with an array of transducers. Section IV depicts sensor's outputs and procedures of validation and comparison against the gold standard for RR in addition to the PPG sensor as another method. Comprehensive data are collected to examine the sensor on six different healthy male subjects in different body kinematic situations such as quiescence, upper body daily motions, big motions and body stretching. Finally, we discuss the implications of our work in Section V.

II. Principles and System Evaluation Method

A. Sensor Position

In this study, our goal is investigating the motion of the diaphragm. So, the sensor was placed on the zone of apposition (ZOA). It is attached to the skin using an adhesive foam similar to ECG electrodes. However, the sensor could be fixed to the body with a stretchable band. The ZOA is the area of the diaphragm encompassing the cylindrical portion (the part of the muscle shaped like a dome) which corresponds to the portion directly apposed to the inner aspect of the lower rib cage [25]. The sensor is placed on the right side of the body, as it is easier to see the diaphragm through the liver window [26]. Previous sensor with one transducer placed in the gap between the ribs was affected by big skin displacement due to possibility of ultrasound wave blockage by rib cages [16]. So, we have arranged three independent transducers in a triangular format in the improved sensor. According to the methods described in the literature [23], [27], [28], the sensor covers ribs 8^{th} and 9^{th} and the gap between them at the midaxillary line, shown in Fig. 1. In this configuration, it can be guaranteed that at least one transducer will always observe diaphragm motions through a rib cage gap. The diaphragm has an hyperechoic structure, which lets ultrasound waves reflected and measured with higher intensity than other surrounding tissues [26]. Normal diaphragm operation in this approach shows the diaphragm moving closer to the transducer with inhalation.

B. Data validation and processing methods

We used a SPR-BTA spirometer as our oral breathing reference for ultrasound data validation. This device is a flow meter not the volume. Therefore, the signal level returns to zero in breath holds. Trapezoidal numerical integration is used to acquire the volumetric values from airflow measurements. The proposed ultrasonic system is based on the volume of air the subject inhales or exhales, not the flow. The spirometer data is logged into computer through GO!Link data logger. A nose clip is used to prevent nasal breathing. The output of this sensor is filtered with a low-pass FIR filter to have a smooth respiratory signal, shown in Figures 4 to 7 as SPR.

Photoplethysmography (PPG) sensor is the most known sensor for heart cycle and oxygen level monitoring. As discussed in the introduction, some researchers utilized this sensor for breath monitoring. We employed a MAX30101 module consists of a PPG sensor and data logger to compare the PPG sensor performance against the reference and proposed ultrasound method. This module has an embedded accelerometer sensor which is used in this research to evaluate the impact of PPG sensor motions on its performance. In all experiments, the PPG sensor was fixed on the finger-tip of subjects to make sure there is no sensor displacement. We focused on the red light of the PPG sensor since we found

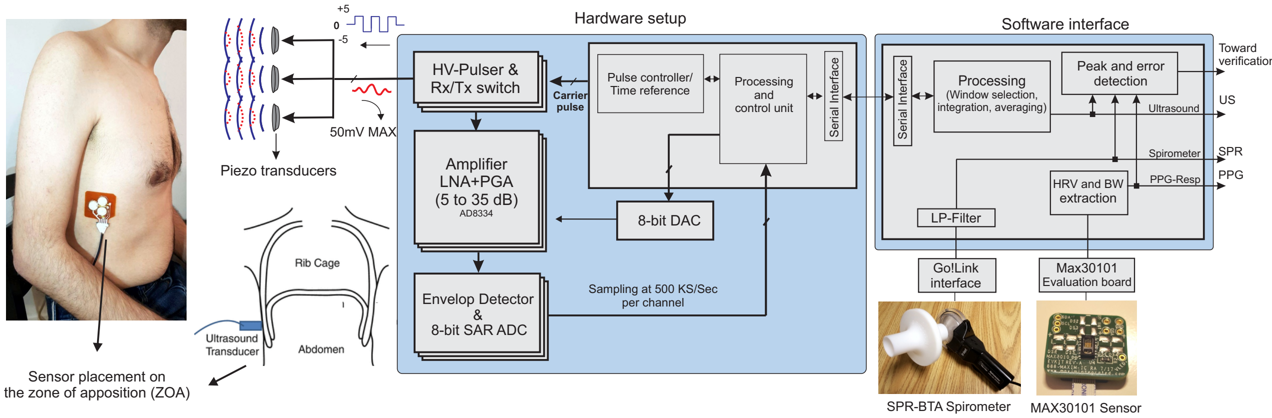

Fig. 1. A) Piezo sensor placement on the body mounted by an adhesive pad. B) Ultrasound system designed to measure and monitor diaphragm motions as a respiratory monitoring system (US-RESP). The data is compared with spirometer, inertial and Photoplethysmography (PPG) sensors.

that sensor's response to the red LED for the heart and respiratory rates are better than the infrared light. The respiratory waveform can be extracted from the PPG via multiple modulations. Two of the most common methods are heart rate variation (HRV) and baseline wander (BW) analysis [29]. A band-pass filter removes a very low and high frequency component of the PPG signal which are results of motion artifact and heart rate, respectively. The resulting is the PPG base-line signal which is in relation with the respiratory signal. Respiratory sinus arrhythmia (RSA) is a phenomenon by which the beat to beat heart rate interval is shortened during inhalation and lengthened during exhalation [30]. So, peak to peak time intervals of heart rate signal for each subject is obtained to form a heart rate variation (HRV) signal. It is expected that the trend of this signal corresponds to the respiratory signal. Changes in venous pressure due to changes in intra-thoracic pressure during the breathing operation cause the baseline modulation of the PPG signal. We have used these two signals to monitor the breathing with PPG sensor. These signals are plotted as PPG-HRV and PPG-Baseline in Figures 4-6.

Another accelerometer sensor is attached to the chest of subjects to monitor the chest motions during the breathing as well as body motions. These motions are shown as the pitch and roll values. All the algorithms and filters are implemented in MATLAB for processing and evaluation.

III. Ultrasound System Description

The piezo disk transducers employed in this work have electrodes on the two faces. One side of the sensor has to be interfaced to a acoustic impedance matching layer with the overall thickness of $\lambda/4$ (λ is wavelength), or less, while having electrical connectivity to the circuit. The other side of the transducer can be covered by a backing material to reduce the transducer ringing effect. In this work, we used a 125 µm thick Kapton (polyimide) film as the sensor substrate. Silver conductive traces (flexible silver conductive ink) are printed on this film for the transducers electrode connection to the circuit, using VOLTERA PCB circuit printer. A conductive silver epoxy is used to mount the transducer on the printed circuit. Since silver is the main component of epoxy, printed ink and transducers electrode, this results in a good acoustic impedance matching between the transducer and the Kapton. Kapton as a flexible and resistant film is a proper substrate that can be easily used as the acoustic impedance matching layer between the piezo and skin. An advantage of impedance matching is the increased sensitivity of the sensor in sensing reflections. To remove the air gap between the sensor and skin, placing ultrasound gel or any similar soft gel is needed [31]. However, a dry-coupled matching layer could be utilized to lessen the need to keep the ultrasound gel. The back of transducers are covered by a hot melt adhesive to prevent excessive vibrations (the ringing effect) and increase the axial resolution. Fig. 1-A shows the designed 3-channel sensor utilized in this research. All the three transducers are excited at the same time to make sure that reflections are met at the same time as well.

The ultrasound system contains three parallel channels of pulser and analog front end (AFE). An 8-channel digital pulser (MAX14808) excites the transducers at a desired voltage. In this study we applied ± 5 differential pulses. This IC has integrated transmit/receive (Tx/Rx) switches as well as grass-clipping diodes for isolations. An AD8334 analog front end magnifies reflected ultrasound waves, as shown in Fig. 1B. So, we can observe a uniform amplitude of signals reflected from the diaphragm wall. Each pulser in this system is controlled by a digital circuit providing a 2.2 MHz carrier signal applied on the PZT5 piezo transducers, for 4 µsec duration. Once the pulse generation ends, the analog switch turns into Rx path to amplify and record reflections. In this design, burst repetition is done 20 times per second, meaning the same rate for every new reflection. Each reflection, called *Records* in this work, contains information about the observing organ position. Since the ultrasound wave attenuates linearly as it penetrates into the body, the reflected signals is amplified with the same rate to compensate the attenuation. So, a low noise amplifier (LNA) together with a programmable gain amplifier (PGA) magnify reflections in this manner. The amplitude and time of flight (TOF) of reflected signals are the valuable information that needs to be captured and processed. So, the envelope of amplified signals is sampled at 500 kilo samples per second, for each channel. We employed a STM32F7 microcontroller to record and process the data. Finally, the captured data are transmitted to a computer for post processing, validation with the reference and comparison with the other methods.

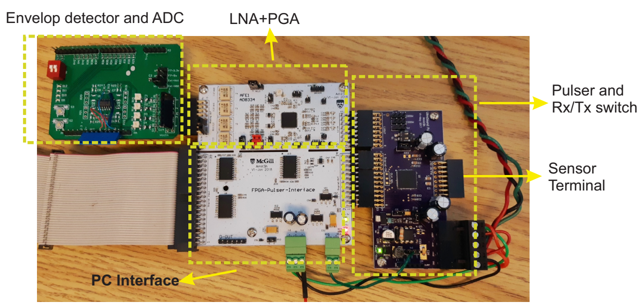

Fig. 2. The designed 4-channel mixed signal hardware setup

The hardware setup of the proposed system architecture is shown in Fig. 2. The system applies differential pulses of maximum ± 8 V on transducers for excitation. The hardware was built using off-the-shelf 4-channel pulser and transceiver integrated circuits (IC). In this work, 3 out of 4 channels are used. This is just an evaluation design and it could be integrated into a small wearable design. The processing of ultrasound data could be done completely on the microcontroller. However, to make a better timing comparison between all three resources, processing was done on the computer for the purposes of our experiments.

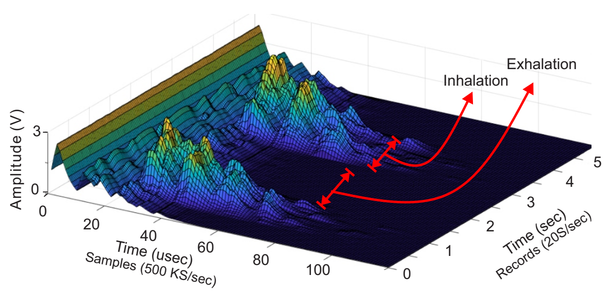

Fig. 3. Five seconds data from ultrasound system when the subject inhales and exhales normally. Signals after the 20 µsec are the results of acoustic reflections from diaphragm wall motions.

Five seconds out of two minutes data (US signal in Fig. 4) from one transducer is plotted in the Fig. 3. In this figure, signals within 0 to 20 µsec are results of ringing effect and skin surface reflections. Signals later than 20 µsec are ultrasound waves reflected from internal organs. The period from 20 to almost 80 µsec on the Samples axis is called *desired window* (DW). An average of all samples within the DW of each record j results the value M_j. A series of M_j values represents the motion of the organ over the time, which is our respiratory raw waveform. The following equation is used to find M_j for all channels (k):

$$M_j = \frac{1}{UB - LB} \sum_{k=1}^{3} \sum_{i=LB}^{UB} S_{k_i} \tag{1}$$

where i is the index of sample S for channel k within the DW period. The LB and UB values are the Lower Bound and Upper Bound of the DW period, respectively. A low-pass equiripple FIR filter is designed to filter some high frequency elements of the M_j raw signal. The filtered respiratory signals are plotted in the Figures 4 to 7 as US. An advantage of the proposed ultrasound system is a simplicity in signal processing. The signal level is always at low value (almost zero) as there is no reflection within the *desired window*. And, the level of signal increases with the presence of diaphragm wall as the reflector of ultrasound waves during the inhalation. So, there is no need to apply extensive signal processing algorithms to subtract the desired signal from the raw data. All these signals are processed in MATLAB in this evaluation stage. However, a simple ARM microcontroller was able to process the data from the 3-channel system at the sampling rate of up to 15 samples per second.

Since all transducers are tracking the diaphragm muscle excursion, ultrasound waves from this organ will be reflected at almost the same time to all transducers. Also, any skin artifact will not impact on signals in the DW period which is a distinct advantage of our system. There is no crosstalk between transducers and overlapping of respiratory and non-respiratory signals observed in different channels. Moreover, reflections from nearby bones have no impact on the signal level within the DW period.

IV. Experiments and Analysis

A. Study Protocol

As described, we are monitoring the internal organ motion, i.e., the diaphragm wall. Our goal is the evaluation of the proposed system on subjects in different body motion and position conditions, including sleep positions. Previously, we observed that the sensitivity of ultrasound sensor in breath detection decreased when there was a sensor displacement due to the body motion or hand abduction [16]. So, in this study two tests are practiced that include these motions. This is intended to examine the sensitivity of PPG, ultrasound and inertial sensors to the motion artifact. Experiments are detailed as the following:

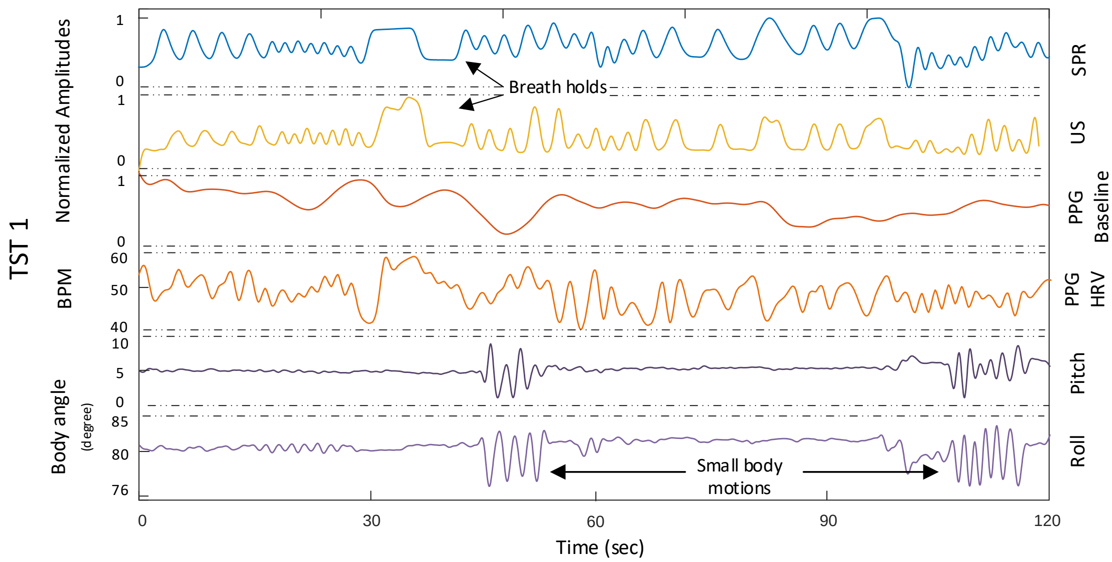

Fig. 4. Respiratory waveforms of a subject as following sequence: spirometer (SPR) as the reference, proposed ultrasound system (US), Photoplethysmography (PPG) and inertial sensor (pitch and roll) to monitor body motions. Note that the intensity of motions are lower than TST2 and TST3.

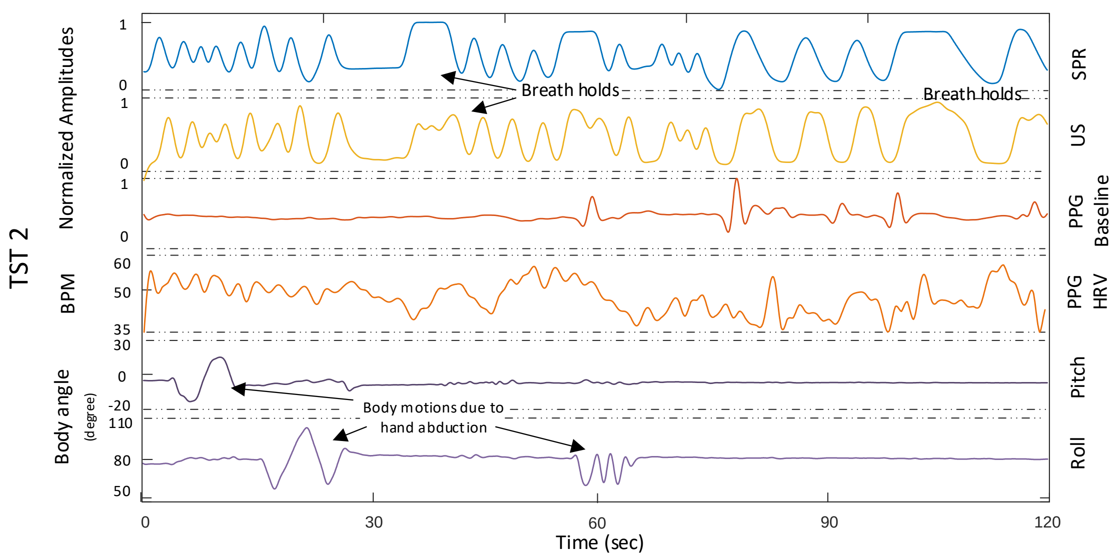

Fig. 5. Respiratory waveforms of a subject as following sequence: spirometer (SPR) as the reference, proposed ultrasound system (US), Photoplethysmography (PPG) and inertial sensor (pitch and roll) to monitor body motions during hand abduction and adduction.

- Test-1 (T1): In this test the subject is asked to disregard the apparatus and breath normally, continue by a breath-hold and then normal and abnormal breathing. This test could be an instance of a suffocation or having difficulties in breathing. The length of breath holding depends on the comfortableness of subjects and is typically between 5 to 30 sec.
- Test-2 (T2): Breathing normally while abducting hands up and down. This test is intended to evaluate the skin movements when the subject abduct (moves away from the body) his arms straight out at the shoulders and then completely up. Thereafter, the subject adducts (moves towards the body) his arms down to the normal condition. It is expected that at least one out of three transducers track the diaphragm motions while there is such a big skin stretches and sensor displacement.
- Test-3 (T3): Breathing with random upper body motion and rotation. In this test the subject is asked to move his upper body randomly in all directions. This is to evaluate the ability of the apparatus when upper body motion is occurring. In contrast with the first two tests in which the PPG sensor was steady, the subject shakes his hand normally to examine the performance of the PPG sensor during hand motions.

- Tests T4 to T6 : One minute normal breathing in different sitting and sleeping positions. Researches have found that the diaphragmatic excursion (DE) varies in different body positions, such as sitting and supine [32], [33]. Takazakura *et al.* found that the DEs in the supine position were significantly greater than those in the sitting position. So, we evaluate the performance of our system in sitting position (T4), supine position (T5) and in left lateral sleep position (T6).

TABLE I
SUMMARY OF SUBJECTS PHYSICAL SPECIFICATIONS. CC IS THE CHEST CIRCUMFERENCE

Subject ID	BMI (kg/m^2)	Age (*year*)	CC (*cm*)
1	24.5	34	88
2	20.7	30	83
3	19.8	29	63
4	27.4	30	92
5	24.8	29	90
6	22.4	37	88

In this study, the experiments were conducted on six healthy subjects aged from 29 to 37, as listed in the Table I. They were instructed how to perform each breathing exercise before their recording sessions. We found oral breathing exhausting if it lasted longer than 2 minutes. So, we set the length of tests to a maximum of 2 minutes.

B. Graphical representation of respiratory signals

Signals of T1-T3 tests for a subject are plotted in figures 4 to 6 and T4-T6 tests are merged into Fig. 7. SPR signal in these figures is the spirometer signal used as our respiratory signal reference for ultrasound (US) and PPG signals. The inertial sensor monitors both the respiratory signal and body motions. The subject body motions are plotted as pitch and roll angles and PPG motions are denoted as PPG pitch and roll.

In Fig. 4, the subject started with normal and fast breathing continued by a full inhalation and exhalation breath holds. The breath holding simulation is intended to evaluate how sensors detect breath pauses in disease such as apnea or Biot's breathing. The subject continued normal and abnormal breathing patterns to the end of this test. It is clear in the Fig. 4 that the US signal is in a high correlation with the reference signal, while the PPG baseline signal has very weak correlation. Heart rate variation (HRV) analysis of PPG sensor is in a high accordance with the reference signal, except the breath holding period. Note that the high amplitude signals of pitch and roll in this figure (around times 55 and 115 seconds) are small motions of the body, not the respiratory motions of the chest.

Fig. 5 depicts the respiratory waveform of all sensors when the subject abducts and adducts his right hand. As explained before, skin displacements versus the rib cage could be the main bottleneck in using ultrasound for internal motion detection. This problem is simply resolved in this article by a three transducer sensor model. The ultrasound sensor never produced false breathing detection. Due to the hand and minor body motions in this test, the PPG sensor failed in detecting correct breathing operation either from baseline amplitude or HRV variations. In this test, because of the body motions, the error rate of inertial sensor in detecting RR increased, as reported in the Table II for TST2.

An example of motion based experiments is plotted in Fig. 6. In this test, the subject bent to four directions by almost 30 degrees for a minute. So, the system experienced sensor displacements in four different directions. Except for the three failures in breath detection at times 32, 50 and 55 seconds, the sensor had a successful operation all the time. The subject continued breathing while motioning his fingers, hand and body in random directions. The ultrasound system could detect all the breathing and non-breathing periods whilst the PPG sensor showed uncertainty in this experiment from time 70 to 120 seconds. The PPG signal amplitude got changed due to its motions as evident in the PPG motion signals. Along the whole test, the PPG sensor was fixed on the fingertip. It is obvious that the inertial sensor can not distinguish minor respiratory motions of the chest from the body motions. Obtained values from TST3 of inertial sensor in Table II imply this fact.

Fig. 7 illustrates all the US and SPR signals of tests 4 to 6. Since the main objective of these tests is to evaluate the performance of proposed system in different siting and sleep positions (as discussed in Section IV.A), the PPG and inertial sensors analysis are skipped in these experiments.

V. RESULTS AND DISCUSSION

We have depicted the respiration detection ability of the proposed ultrasound sensor as well as PPG and inertial sensors as two conventional methods introduced by other researchers. This section evaluates all the methods statistically for all subjects and tests.

Table II reports the evaluation of the T1 to T3 tests results on all six subjects and Table V reports the later T4 to T6 tests. We have used windowing percentile (of about 80%) analysis on all data to determine peaks and valleys. The resulting would be a square signal that the two levels are representations of the inhalation and exhalation cycles. This method is independent of the signal offset (baseline). However, even if there is no object (diaphragm) reflecting ultrasound waves, the amplitude of the signal is near to zero. So, any amplitude higher than a tunable threshold would be a certainly true inhalation. This is an advantage of this sensor for providing accurate data and simple processing effort. The percentile score is relative to the size of the window. A window size of 10 seconds and a dynamic percentile score of 60% to 80% were ideal combinations for this aspect. Please note that in this work, the goal is to identify every inhalation and exhalation cycles (*i.e.* a full breathing operation). So, every peak and valley of the

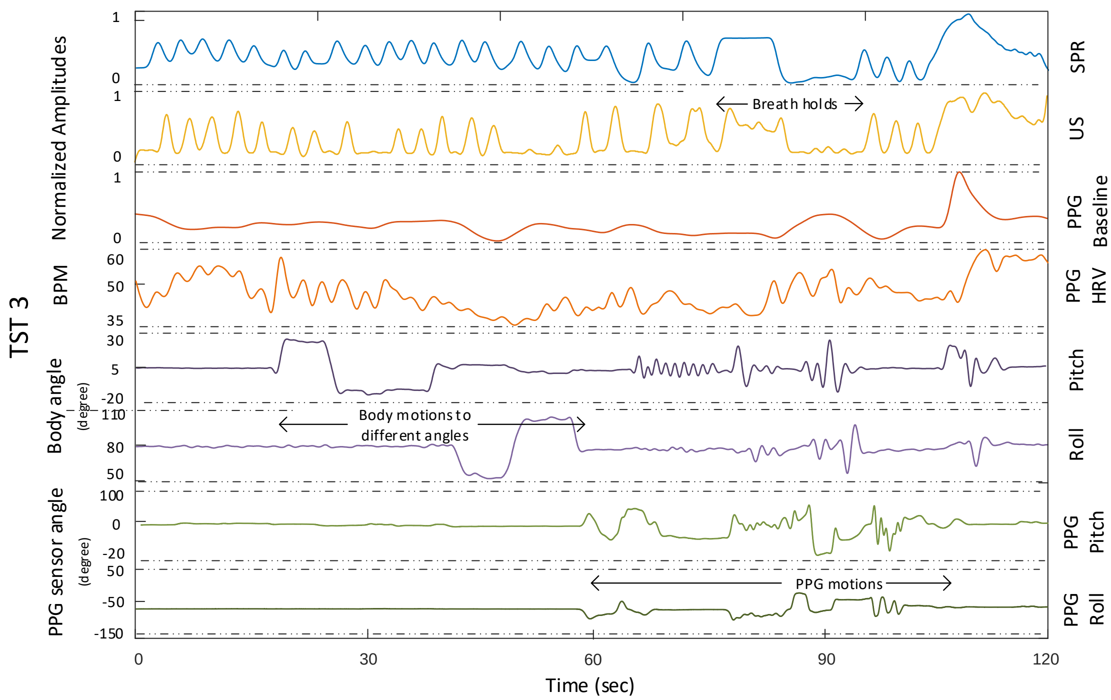

Fig. 6. Respiratory waveforms of a subject as the following sequence: spirometer (SPR) as the reference, proposed ultrasound system (US), Photoplethysmography (PPG) and its motion on the finger-tip, inertial sensor (Gyroscope and Accelerometer) to monitor body motions.

TABLE II
SUMMARY OF STATISTICAL MEASURES OF THE PERFORMANCE ON SIX SUBJECTS FOR TESTS TST1 TO TST3. ALL VALUES ARE IN PERCENTAGE.

Test number	Average breaths /length(min)	Ultrasound				Inertial Sensor				PPG-HRV				PPG-Base			
		Sens	Spec	Prec	FAL	Sens	Spec	Prec	FAL	Sens	Spec	Prec	FAL	Sens	Spec	Prec	FAL
TST 1	22	89.5	93.5	94.3	6.4	76.2	86.8	82.8	13.1	84.6	84.2	83.8	15.6	55	77.6	66.4	22.7
TST 2	21.8	92	94.7	94.8	5.2	50.6	80.2	67	19.9	86.4	91.2	90.8	8.8	33.2	79	58.4	20.9
TST 3	20.9	87.7	91.2	90.8	7.3	17.7	59.7	31	40.4	82	86.3	84.3	13.8	44.5	74.2	64	25.7
Average	21.6	89.7	93.1	93.3	6.3	48.2	75.6	60.3	24.5	84.3	87.2	86.3	12.7	44.2	76.9	62.9	23.1

SPR signal as a reference is compared with the corresponding peak and valley position of the signal from another sensor such as US or PPG. The total number of true and false detected events for the existing and non-existing breath operations are counted to calculate the sensitivity (true positive rate (TPR)), specificity (true negative rate (SPC)) and precision (positive predictive value (PPV)) values of the three methods regarding to our reference. False alarm ratio is added to this table to compare the probability of generating false positive patterns in different sensors. For applications such as sleep apnoea, false breathing detection from a suffocated patient would lead to irreparable harms such as hypertension, heart disease including heart attacks and heart failure, diabetes and stroke.

According to the average values (average of 6 subjects in each test) in the Table II, the proposed ultrasound sensor scored better than the PPG and inertial sensors for all sensitivity, specificity and precision values with about 90% and more. It also ranked first in generating the least false detections. In the previous work, the average of obtained sensitivity and specificity values were 73% and 89% respectively, for tests that involved body motions [16]. In this work, outcomes of Test3 were 89.7% and 93.1% for the sensitivity and specificity parameters.

Moreover, based on the Table III and Fig. 7, the ultrasound sensor had high sensitivity, specificity and precision for all tests 4 to 6. As mentioned before, researchers showed that the diaphragmatic excursion (DE) varies in different body positions [32], [33]. They found the DE is greater in supine position than in the sitting. Based on our finding in this work, we did not perceive any perceptible changes in respiratory signal and system performance due to the DE variation in different positions.

The reader must note that the relationship and magnitude of RSA has been shown to be easily influenced and diminished by poor cardiopulmonary function and disease such as coronary artery disease. In elderly people, the abnormal RSA is very prominent, while the importance of respiratory monitoring is higher for them. Poor physical fitness is another instance in which for both

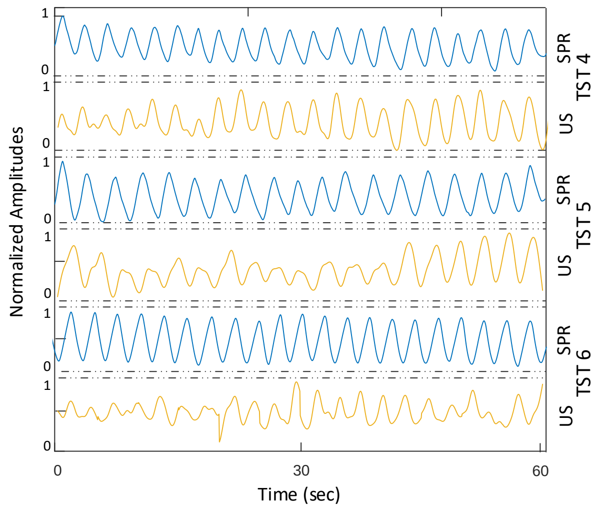

Fig. 7. Respiratory waveforms of a subject in sitting (TST4) position, supine (TST5) and left lateral (TST6) sleep positions.

TABLE III
SUMMARY OF STATISTICAL MEASURES ON SIX SUBJECTS FOR TESTS TST4 TO TST6.

Test number	Ultrasound			
	Sens	Spec	Prec	FAL
TST 4	95.4	96.6	97.2	3.3
TST 5	87.8	96.8	96.6	3.2
TST 6	90.2	92.2	91.4	7.7
Average	91.1	95.2	95.1	4.7

athletes and people who routinely exercise have higher prevalence of RSA than people who do not exercise [30] [34]. These reasons may exclude the PPG sensor as an applicable sensor for respiratory monitoring. Although the HRV analysis on the PPG sensor yields an acceptable accuracy (see Table II), this sensor is still an indirect method for respiratory monitoring.

VI. CONCLUSION

In this paper, we evaluated a multichannel pulsed ultrasound system (US) for monitoring respiratory cycles in different human body motions as instances of daily activities. The system utilizes three PZT5 piezo transducers, mounted on a flexible surface, working in parallel to track the diaphragm motions observed from the zone of apposition (ZOA). The proposed sensor probes internal organ motion (the diaphragm muscle), rather than the external or surface motions of the body, making that extremely independent of patient's movements. The system output was examined by a spirometer as the gold standard. Inertial and photoplethysmography (PPG) sensors were employed to evaluate their operations as two alternative methods in comparison with the proposed system. The ultrasound system exhibits better performance than the PPG and inertial sensors under the body motions.

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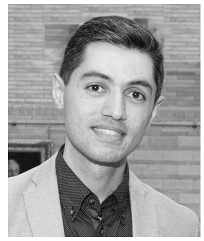

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