A Novel Respiration Rate Monitoring System Using Optical Technique

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Abstract— The respiration rate of patients is an important vital sign that can be used in diagnosing various diseases such as chronic obstructive pulmonary disease, lung cancer…etc. It must be continuously monitored during any surgery and in the post-anesthesia care unit. A nasal cannula or a face mask Capnometry is the standard tool for monitoring the respiration rate. However, such tools might be poorly tolerated and displaced, leading to inaccurate measurements or false alarms. In this paper, a novel noninvasive system is presented to measure the respiration rate of patients. The system is based on a LED transmitter and an optical detector array. The results show that the sensory system is able to follow and record the breathing rhythm perfectly. The amplitude of the measured signal and number of peaks indicate the respiration depth and rate, respectively. The number of pulses at the output of the system matches very well the counting of the respiration rate taken by the subject. This system has an advantage in performing the respiration rate measurements non-invasively to avoid the problems of dislodging or displacing the sensor at the interface. Additionally, it has an advantage of separating the sensor from being in-contact with the subject body, which is an important factor to avoid the risk of infectious diseases. Such a system can be considered as a cost effective solution for noninvasive respiration rate monitoring systems.

Keywords— Breathing monitoring, Optical sensing, Respiratory rate.

I. INTRODUCTION

Respiration is the process of gas exchange throughout which oxygen and carbon dioxide are transported and exchanged between the lung and external environment. The inhalation and exhalation of air allows the gas exchange to occur within the alveoli in the lung due to diffusion [1]. The outcomes of this process are the bonding of oxygen molecules with the blood and the excretion of carbon dioxide, which gets out of the body through the nose. This process occurs frequently in a cyclic form (i.e. respiration cycle). The number of cycles per minute is called the respiration rate.

The respiration rate varies according to the personal needs for oxygen; and it is a vital sign for individual health. Many studies showed that the alteration of respiration rate can be used to predict potentially serious clinical events and monitor the progression of serious illnesses [2]-[4] such as predicting influenza, asthma, tuberculosis, lung cancer and chronic obstructive pulmonary. Some medical views recommend the use of both pulse oximetry (SpO_2) and respiration rate monitoring to assess medical conditions. After extubation, patients receive supplemental oxygen. SpO_2 only can be a belatedly indicator of alveolar hypoventilation [5]-[6].

Recent research in respiration rate monitoring has rapidly developed the detection of both the chest and abdominal movement and the exhausted gases properties (i.e. temperature, flow etc.) to attain a reliable method of measurements [7]-[8]. Different techniques have been used to measure the respiration rate, which can be classified as direct or indirect methods.

In the direct methods, special types of sensors are positioned directly onto and in contact with the patient body (ex. optical sensors, ultrasound sensors…etc.) [9]. Since it is directly coupled to the patient’s body, it provides accurate measurements and improves patient comfort.

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The indirect methods are performed by extracting the respiration signal embedded in another bio-signal(s). Examples of the indirect method include processing the ECG signal to extract the respiration rate signal and using blood gases measurements to assess respiratory activity [6]-[9].

In literature, many methods have been described to monitor the respiration rate; mainly the acoustic and nasal airflow-sensing, chest and abdominal movement detection and transcutaneous CO_2. The acoustic based method measures the breathing sound at the throat using a microphone. The drawback of this method is the possibility of sounds interferences during the measurements (e.g. snorting, speaking, crying, and coughing). This makes it unsuitable for long-term or comfortable use. Nasal Airflow-sensing method measures the variation in temperature in the exhaled and inhaled air. However, the use of this method is limited due to the high incidents of sensors. Chest and abdominal movement detection method is based on measuring the wall movement of the chest by a strain gauge placed on the rib cage. The abnormality in the lung function causes asynchronous movement of the chest; and hence, the system could fail to provide accurate measurements [10].

The respiratory rate can be also estimated indirectly by processing other biosignals such as the impedance pneumography (IP) signal across the chest, ECG signal and Photoplethysmogram (PPG) signal from finger tips [10]-[13]. However, motion artifact can disturb the signal acquisition and thereby deteriorate the reliability and accuracy of measurements. Comprehensively, the review papers by Folke *et al.* [14] and Al-Khalidi *et al.* [15] have addressed most of the known respiration rate monitoring methods in literature.

The normal respiratory rate can be classified according to the ages of individuals as shown in Table 1, which shows that the respiratory rate decreases with age. Children have higher rates than the elderly while males have lower respiration rates than females.

TABLE 1
AGE GROUPS VERSUS RESPIRATION RATE [16]-[18]

Age Groups	Respiration Rate
Newborn	30-60 bpm
Infant (1-12 months)	30-60 bpm
Toddler (1-2 years)	24-40 bpm
Preschooler (3-5 years)	22-34 bpm
School-age child (6-12 years)	18-30 bpm
Adolescent (13-17 years)	12-16 bpm
Adult	12-18 bpm

A change in the respiration rate can be used as an indication to different diseases such as laryngitis, bronchitis, tonsillitis, pneumonia, bronchiectasis, influenza, asthma, tuberculosis, lung cancer, chronic obstructive pulmonary disease, emphysema, and pulmonary embolism [19]. This theme will be investigated in the future research.

Recently, noninvasive measurement techniques in biomedical applications are evolving rapidly due to their advantages over invasive measurement techniques. Performing the measurements on living organisms without breaking the skin is the greatest advantage of such systems [20]. Using an optical light to perform these measurements on the human body draws information and adds another value regarding safety and health. In this paper, we will address a novel method to noninvasively monitor the respiration rate. The presented technique is based on detecting the chest movement due to the respiratory activity using optical sensors.

II. SYSTEM DESIGN:

We have developed a novel system for measuring the respiratory rate (see Fig. 1). This system is composed of a light source and a photo-detector array to detect the changes in the reflected light during breathing. The sensor is fastened with a belt on the human chest onto a barrier (i.e. white underwear) to follow the abdominal and chest volume changes caused by breathing. Signal acquisition is used to interface the sensor with a computer for further processing. Such sensors can be considered as the base for future developments to produce inexpensive, user friendly device that can be routinely used in clinics or at home.

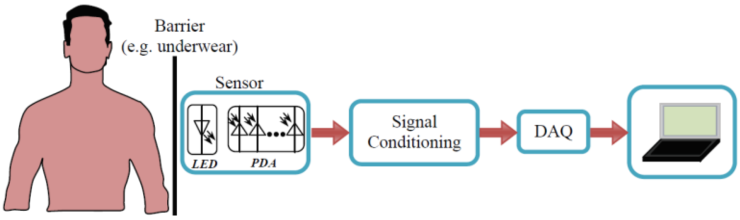


Fig. 1. Block diagram of the proposed system

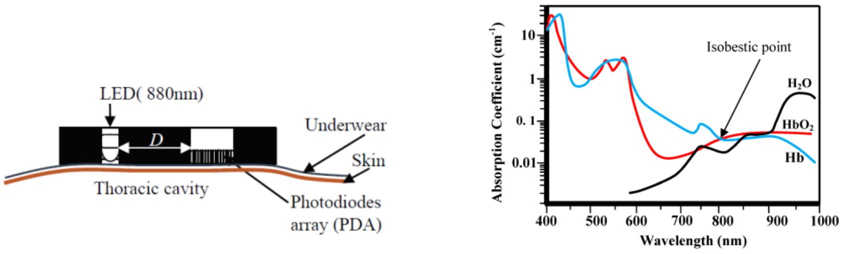
III. SENSOR DESIGN AND PRINCIPLE

The sensor attached to the subject body is a reflection type sensor. Fig. 2a shows a cross-section view of the developed sensor used in this study. A near-infrared (NIR) LED with 880 nm wavelength (OPE5T87 model developed and manufactured by Roithner Lasertechnik) is used as an optical light source. The wavelength is selected in such that the absorption coefficient of oxygenated and deoxygenated blood is low to increase the amount of the reflected light. Additionally, reflections due to underwear or different tissue constituents are smaller than the other wavelengths in the optical therapeutic window. The 880 nm wavelength is selected in the ranges following the isobestic wavelength, in which the absorption coefficient of oxygenated and de-oxygenated blood is almost identical (see Fig. 2b). Thus, the sensitivity of the sensor increases due to the increase in the optical density of the reflected NIR light. The thoracic volume change reflects the major changes in the signal components. The backscattered light from the thoracic cavity is detected by a linear array of photodiodes (photo-detectors TSL1401CL model developed and manufactured by AMS-TAOS), which have been optimized to have the maximum sensitivity at the desired wavelength (i.e. 880 nm). The photodiodes array consists of 128 elements integrated on a single chip and separated from the light source by a distance *D* that equals 12 mm. The resulted signal in response to the intensity of the light source is proportional to the amount of charges accumulated in the integrated charge amplifier circuit within the detector array. The measured signals are controlled by shift register which holds it and sends it to the output stage through an op-amp. The resulted signal contains high frequency components (i.e. in kHz ranges as originated from the multiplexing process) that can be digitally filtered. It is considerably higher than the frequencies of the signal of interest (i.e. 0.05 Hz to 10 Hz).

The light transmitted by the NIR LED is diffused through the chest via underwear and skin layers. During this process, the diffused light interacts with the chest and different tissue constituents which alter its intensity. The interaction involves reflection, refraction and absorption. Although the interaction of light is complex, simple model like Beer-Lambert law (BLL) is applicable here. The optical density of the back scattered light represents the amount of light being attenuated or absorbed by the medium change; the absorbance A is given by:

$$A = log_{10}(I_o/I) \quad (1)$$

where I_o is the incident intensity; and I is the received intensity.



a *b*

Fig. 2. a) Cross section view of the respiration rate sensor placed on the thoracic cavity, b) Optical window

The amount of the optical density depends on many parameters such as different layers structure, the sender-detector separation *D* and the energy delivered by the LED. The energy of LED is inversely proportional to the wavelength according to:

$$E = h \cdot v = h \cdot (c/\lambda) \quad (2)$$

where E is the energy in Joule; h=6.626×10^{-34} J.s is Planck's constant; c=2.998×10^{8} m/s is the speed of light in a vacuum; and λ is the wavelength in nanometer.

The energy must be sufficient for the light to penetrate through the underwear to the chest. Likewise, the optical output intensity is an important criterion in selecting the LED operating wavelength. A driver circuit (shown in Fig. 3) is used to control the light intensity of the LED.

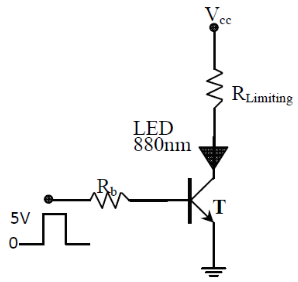


Fig. 3. NIR LED driver circuit

The data acquisition card used in our experiment (i.e. DAQ card from national instrument (NI) model NI USB-6216) has 16 bit resolution with a sample rate up to 400 kS/s.

Sensor arrays consist of 128 photo-detector elements; and the acquisition time of each photo-detector element is limited to 20 µs. Thus, the sampling rate of the system should be 50 kHz. (i.e 50 kS/s= 1/20 µs).

IV. Methodology and Signal Processing:

Using our system, three different respiration patterns were examined, namely paced breathing, non-paced breathing and free breathing at different conditions. While some of the signals were measured at the chest on top of the underwear, others directly on the skin. Each of the measurements was recorded while subjects were seated and asked to breath according to the required patterns. The measurements were performed on male subjects with ages (44-46) years old. All signals were recorded with a sampling frequency of 50 kHz for duration of sixty seconds. The breathing rate has been determined using the reciprocal of the time between two consecutive peaks in the measured signal.

The raw signal acquired by the sensor was corrupted by a high frequency noise due to the time division multiplexing process (i.e. switching noise). Therefore, further processing was performed on signal amplification, separation, wave-shaping and filtration. Some of these signal conditionings were performed by hardware (the signal conditioning block diagram in Fig. 1) while the others were handled in software under LabVIEW environment. Fig. 4 shows an image for the entire system during the measurements; and Fig. 5 shows the signal processing blocks in LabVIEW environment used in the measurements.

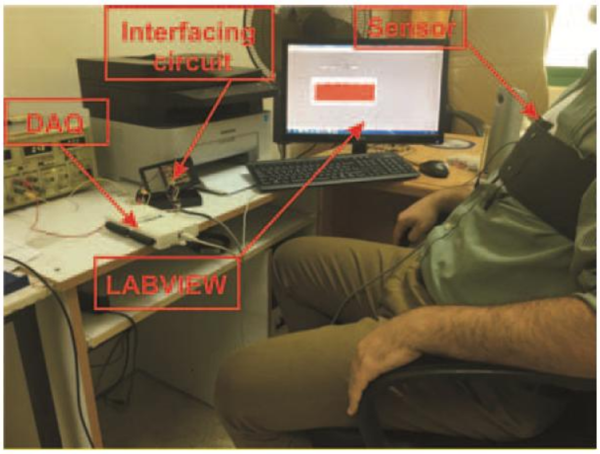


Fig. 4. An image for the system used in the measurements

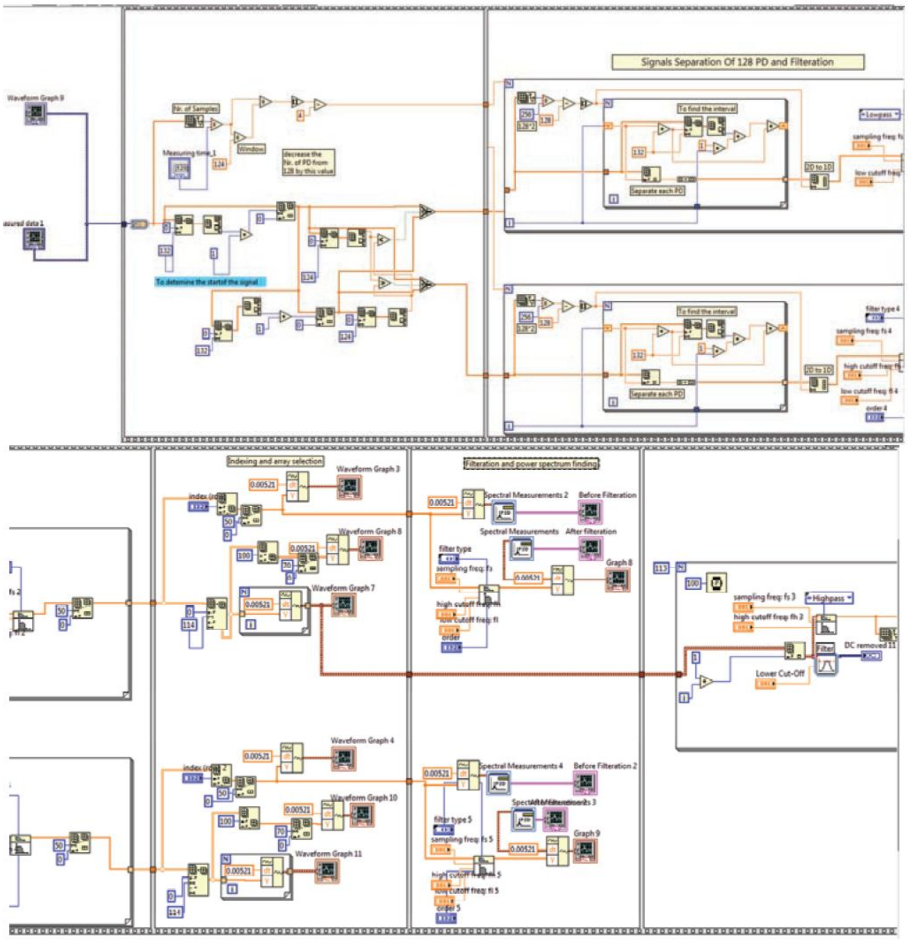


Fig. 5. An image showing the LabVIEW blocks used in the measurements

In all of the measurements performed in this study, the signals were converted to the digital form and processed to determine the main features (i.e. signal peaks and valleys). The 128 signals generated by the photo-detector arrays were separated as they were time multiplexed. Fig. 6 shows an example of measured respiration signals as generated by 128 photo-detector elements.

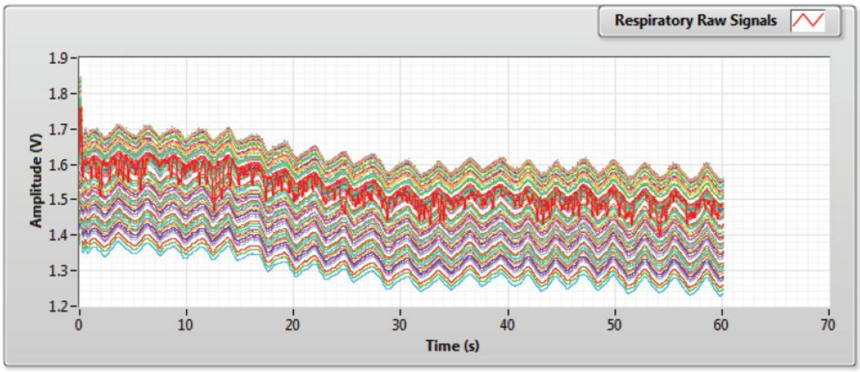
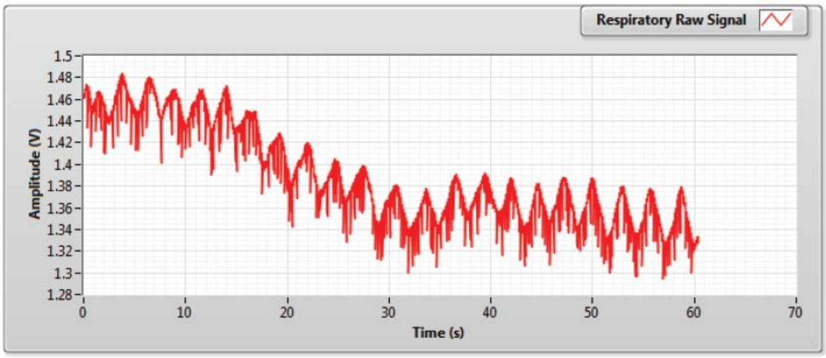


Fig. 6. The de-multiplexed signals as generated by 128 photodiode arrays

Subsequently, the measured signals were digitized with 50 kHz sampling frequency. Afterwards, digitally band-pass signals are filtered to remove power-line interference, high frequency noise and baseline wandering. The corner frequencies of the band-pass filter were chosen to be f_{Lower}= 0.01 Hz and f_{Upper}= 10 Hz. Fig. 7a shows the raw signal of one of the photo-detectors. The filtered version in Fig. 7b and Fig. 8 shows the power spectrum density of the processed signals (i.e. normal breathing rhythm) with the fundamental frequency component at 0.4 Hz.



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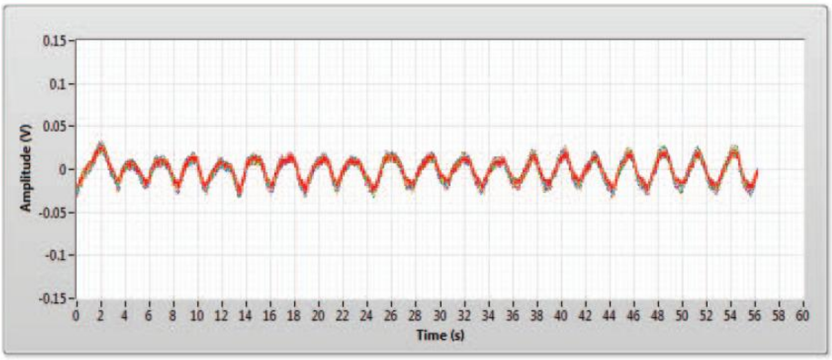


b

Fig. 7. a) A raw signal as detected by one photo-detector, b) Filtered signals of the raw signal

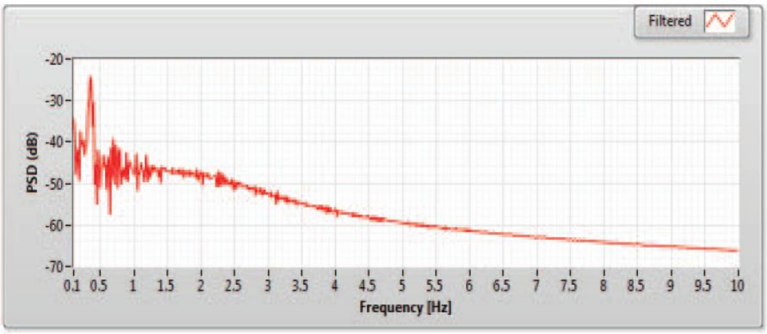


Fig. 8. Power spectrum density of the signal shown in Fig. 7b

V. RESULTS

The results show that the pattern of the generated signal from respiration rate sensors has a pulsatile modulated signal shape. The change in the volume conductor during breathing alters the amount of the backscattered light received at the detector side. Fig. 9 shows an example of a measured respiration rate signal under a paced breathing condition.

In Fig. 9, the amplitude of the pulsatile of the signal was about 56 mV while the breathing rate was 24 bpm. This matches the real breathing counts taken by both the assistant and the subject himself.

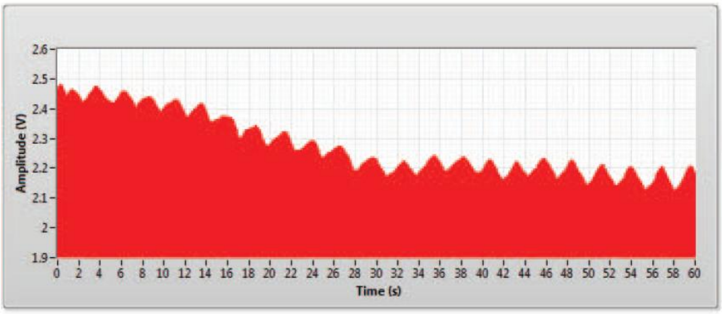


Fig. 9. Paced breathing process

The sensor ability to follow and record the breathing coincident has been investigated under different conditions. The results show an ability of the sensor to follow perfectly well breathing incidents. During paced breathing, the subject was informed to take a deep breath during measurements before turning back to normal breathing. In consequence, the amplitude of the pulsatile of the measured signal has increased to 480 mV in the period 8-22 s. The breathing rate was about 27 bpm (see Fig. 10).

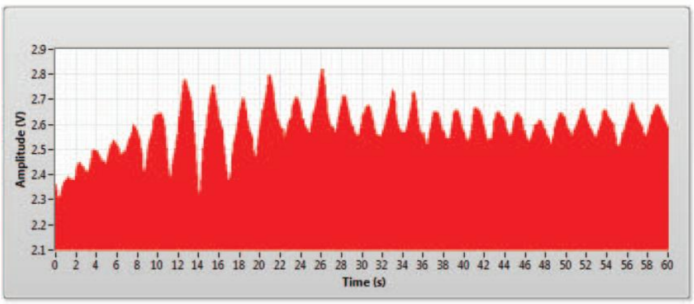
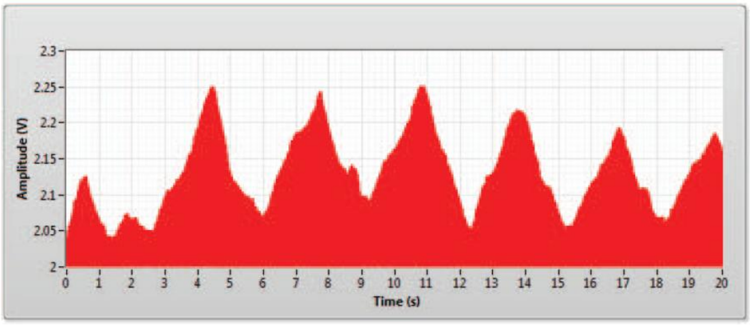


Fig. 10. Mixed rhythm paced breathing with deep breathing

The sensitivity of the system has been also investigated by measuring the deep breathing signal on top of a barrier. The sensor was mounted on top of a white underwear; and the signal was measured for 20s (see Fig. 11a). The system showed a good performance (i.e.

signal amplitude and shape) when it is measured with a barrier. Subsequently, the signal amplitude decreased approximately to the half without affecting the quality of respiration rate measurements (i.e. 215 mV with 21 bpm breathing rate). One advantage of our system is that it performs non-contact measurements and reduces the risk of infection.

The response of the system has been investigated by considering a high breathing rhythm. The results show that the system was able to follow the changes in the breathing rhythm. The signal amplitude was nearly the same as in the previous case (i.e. non-contact paced breathing measurements), so the breathing rate has increased (as expected) to 36 bpm, reflecting the real breathing rhythm (see Fig. 11b).



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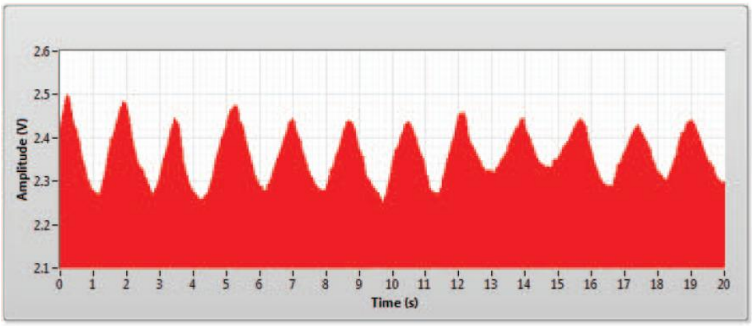

b

Fig. 11. a) A 20-second deep breathing rhythm on top of the underwear, b) A 20-second increased breathing rhythm on top of the underwear

Another experiment was performed by considering deep breathing in a very low rhythm (see Fig. 12). The resulted signal shows ten peaks of the breathe rhythm (i.e. 10 bpm) with a signal amplitude of 280 mV.

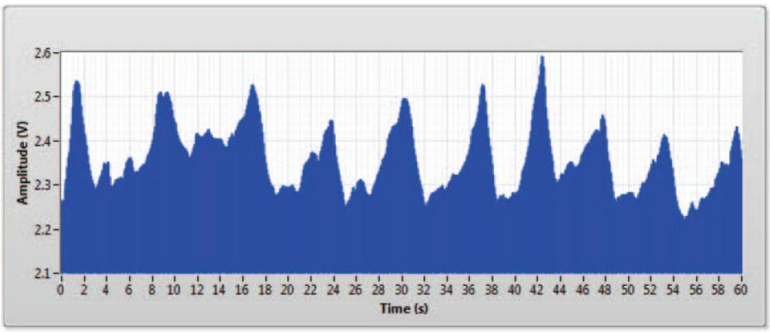

Fig. 12. Deep slow breathing rhythm on top of the underwear

Non-paced variable breathing measurements were performed with deep and normal rhythm breathing conditions. The resulted signal perfectly follows the rhythm of breathing. The deep breathing rhythm is clearly shown in the output signal in the time slots (from 6-14 seconds and from 32-44 seconds) as its amplitude is higher than at other time slots (see Fig. 13). Fig. 14 shows the system results under chaotic breathing conditions.

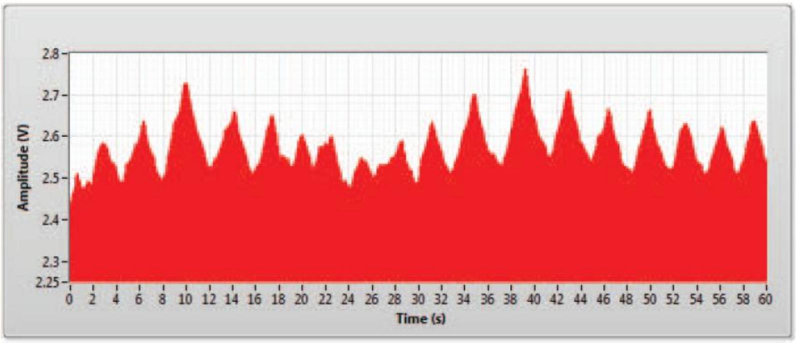

Fig. 13. Variable breathing measurements on top of the underwear

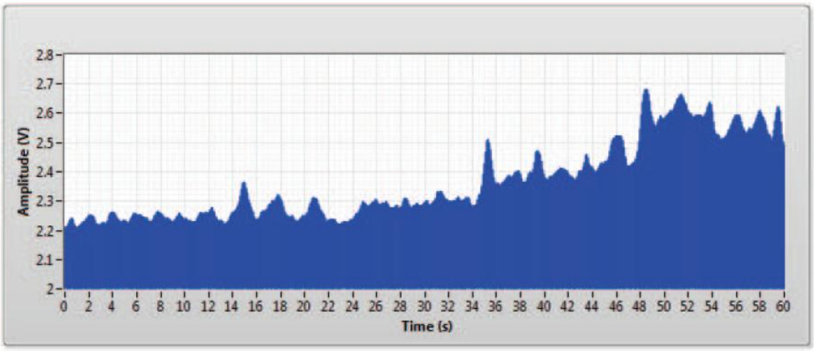

Fig. 14. Chaotic breathing measured on top of the underwear

VI. DISCUSSION

An optoelectronics sensor was designed and utilized to measure the respiration rate. The optical sensor was used to perform different measurements at different conditions. The results addressed above showed the ability of the presented system to track respiration rhythms perfectly well. The parameters used to characterize the performance of the system were both the amplitude and frequency of the measured signals. The amplitude of the signal was used to distinguish between different kinds of breathing (e.g. normal breathing, deep breathing or both combined). The amplitude decreased approximately to the half if the measurements were performed through a barrier (like the underwear) as shown in Fig. 11. This can be related to light losses at sensor-underwear interface. Though the amplitude of the measured signal with underwear deteriorates, the system is still able to measure the respiration rate. This is advantageous for the system in performing non-contact measurements; it also helps avoid disease infection.

The respiration depth can be used as an early sign and symptom for many health events like cardiac arrest [21]. Measurements emphasized the natural relationship between the depth of breathing and respiration rate. As the depth increased, the rate became slower (Fig. 11a). These measurements demonstrate the differences between normal and deeper breathing (higher light intensity, lower frequency) which can be used to monitor the condition of the subject.

Additionally, the detection of the respiration rate and pattern enables medical intervention to avoid the deterioration of a patient’s condition. Such a system can be used in continuous long-term monitoring for the patient in intensive care-unit. Moreover, the designed system

comprises many features like low complexity and ease of use, patient safety, interference free and small size.
The functionality of the system is limited by the light interference with background light. Therefore, the sensor must be perfectly isolated; and the measurements should be performed in a relatively dim room. Attaching the sensor to the chest of the subject is one of the main issues that the operator should consider during performing the measurements.
The respiratory failure of the patient can be identified using our system by considering chaotic breathing conditions in which the subject started to breath in an irregular manner. Under this condition, the signal was unclear and not uniform. Fig. 14 shows that a rapid fluctuation in the amplitude and frequency of the measured signal has been recorded.

VII. CONCLUSIONS

A novel system for measuring the respiration rate using optical techniques was developed, implemented and tested. The system consists of an optical light source with a photodetector array. The system has shown an ability of measuring the respiration rate with and without barriers (i.e. underwear) under different conditions such as paced breathing, non-paced breathing, deep breathing and chaotic breathing. The results show that the respiration rate matches the real breathing counts very well. Non-contact measurements can be considered as an advantage of this system to reduce the danger of infection caused by individual use.
For future work, the system could be improved in both hardware and software. Hardware filtration is recommended in conjunction with post amplification to obtain a more robust free noise signal before digitizing it. Additionally, the system can be used to diagnose some diseases (i.e. respiratory system diseases), which necessitate long-term monitoring.

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