Malaysian Journal of Fundamental and Applied Sciences Vol.9, No.3 (2013) 129-133

ISSN 1823-626X

**Malaysian Journal of Fundamental and Applied Sciences**

**available online at http://mjfas.ibnusina.utm.my**

# Noncontact Respiration Rate Monitoring Based on Sensing Exhaled Air

A.H. Alkali[1], R. Saatchi[1], H. Elphick[2], D. Burke[2] and R. Evans[2]

[1]*Faculty of Art, Computing, Engineering and Sciences, Sheffield Hallam University, Sheffield, United Kingdom*
[2]*Sheffield Children's National Health Service Foundation Trust, Sheffield, UK.*

Received 1 December 2012, Revised 25 March 2013, Accepted 16 April 2013, Available online 2 May 2013

**ABSTRACT**

Respiration rate is the average number of times air is inhaled and exhaled per minute. Respiration rate is an important indicator of a person's health and therefore, it needs to be measured accurately. Existing respiration monitoring systems are generally contact based that means the sensing element needs to be attached to the subject's body. The attached sensor can cause distress in some children, affecting their respiration rate. The device can also become dislodged interrupting the monitoring. This work presents an air flow sensing approach to noncontact respiration rate monitoring. The exhaled air is guided through a small funnel to a chamber that contains a heating element. The heated air leaves the chamber and is then detected by a thermistor that converts the air flow temperature variations to an electrical signal. The signal is amplified, filtered and digitised. Signal processing techniques are used to extract respiration rate from the signal in real time. The device provides respiration rate at distances from 15 to 30 cm from the subject's face.

| Respiration rate monitoring | physiological signal analysis | medical electronics engineering |


http://dx.doi.org/10.11113/mjfas.v9n3.97

## 1. INTRODUCTION

The respiratory system supplies essential oxygen to the human body to allow it function correctly [1]. The average number of breaths taken per minute is known as the respiration rate. Respiration rate depends on factors such as the age and physiological conditions [2]. In healthy subjects it varies from about 15 (in adults) to about 60 (in infants) cycles per minute. The rate needs to be accurately monitored to assess a patient health.

Table 1 summarizes the range of respiratory rates for healthy subjects of different age range.

Table 1: Respiratory Rates in Different Age Groups

| Age Group | General Respiratory Rate |
|---|---|
| New-born and Infants | 30 - 60 |
| Infants | 24 - 30 |
| Toddlers | 20 - 30 |
| Children | 12 - 30 |
| Adults | 08 - 20 |

Respiration process produces a variety of effects that can be used to measure its rate. A review of the main methods is provided in [3]. The methods rely on one of the following effects or signals.

*Chest movements*: Chest and abdominal movements are detectable by a mercury strain gauge or impedance analysis methods. Two bands can be used for this purpose. The thoracic band is placed around the rib cage and the abdominal band is placed over the abdomen at the level of the umbilicus to measure respiration rate [4,5].

*Exhaled and inhaled air temperature variation:* Exhaled air is warmer than inhaled air. A temperature sensitive device such as a thermistor, placed close to the nostril converts these temperature changes to an electrical signal that is then used to determine respiration rate.

*Oximetry Probe (SpO2):* A pulse oximeter uses red and infrared lights to determine the percentage of oxygenated hemoglobin in the blood [6] Respiration rate can be extracted from the signal obtained by an oximeter [7].

*Electrocardiogram (ECG) Derived:* This method is based on the fact that respiration process has a modulating effect on the ECG. By measuring the fluctuation in ECG, the respiration rate can be derived. This technique is called ECG-derived respiration rate monitoring and is based on a process known as sinus arrhythmia, that is, the modulation of ECG by the breathing process [8].

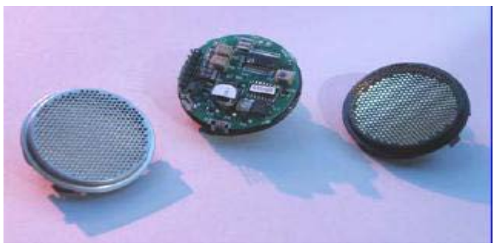
*Exhaled air carbon dioxide ($CO_2$):* Exhaled air contains more $CO_2$ than inhaled air. Thus a$CO_2$ detecting device can covert these changes to a respiration signal [9].

*Infrared emission:* When air is exhaled, the temperature of the skin surface on and around the tip of the nose increases, causing a higher level of infrared emission from the region. These fluctuations can be monitored using a suitable infrared detection device. Signal and image processing techniques are then used to extract a respiration signal and rate [10].

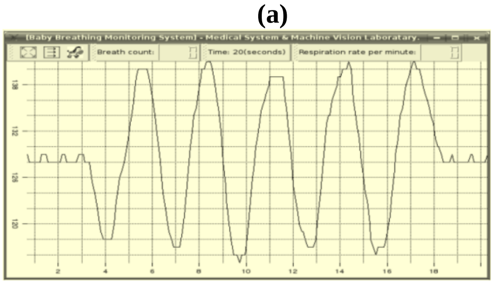
*Corresponding author. E-mail: R.Saatchi@shu.ac.uk

The existing respiration monitoring systems generally require the sensing device to be wired to and attached to the subject's body. The attached sensor causes stress in some children thus affecting their respiration rate. Sensor movements affect the recording and when they are dislodged the monitoring is interrupted. Therefore, there are advantages to monitoring respiration rate at a distance from the subject (i.e. in a noncontact manner).

We have investigated a number of noncontact respiration rate monitoring approaches. Chest movements were detected using an ultrasound sensor. The sensing device used was MINI-A PB ultrasonic transducer [11][9]. This detected chest movements with an accuracy of 0.5 mm.



**(a)**

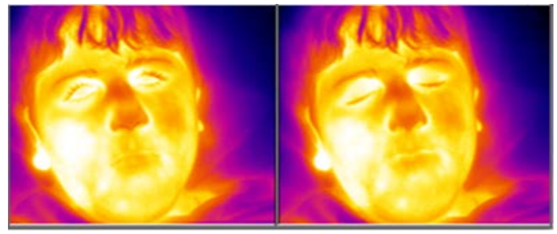



**(b)**

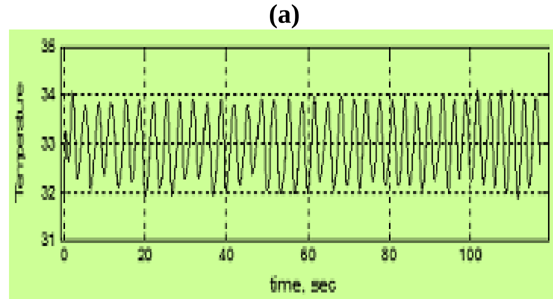
Fig. 1 (a) Senscomp's MINI-A PB ultrasonic transducer, (b) respiration signal generated using the device.

During inhalation, the chest moves forward and then back during exhalation. The ultrasound sensor compares the signal transmitted to the one reflected from the chest to determine its distance relative to the chest. The system works well in controlled situations, where the subject does not move. Body movements affect the chest distance to the sensor, influencing the accuracy of the readings. Also loose clothing interferes with the recording.

Facial thermal imaging is an approach to respiration monitoring [10]. For this purpose an advanced thermal camera was used (FLIR A40). The camera was mounted from about one metre from the subject's face. The infrared emission changes caused by respiration were detected by the thermal camera. A series of thermal images were produced (at 50 images per second). The images were segmented to identify the area around the nose and then a feature from this region plotted against time (i.e. for the successive recorded images) provided a respiration signal. The selected feature was the average of pixel values in the selected area. The pixel values represented the skin surface temperature. Digital signal processing techniques were used to determine the fluctuations in the signal caused by respiration. The approach was evaluated in a hospital environment against a number of conventionally used contact respiration monitoring methods. These included thermistor placed closed to the nostril and pressure sensors to detect chest movement. The thermal imaging method provided results that closely matched the best contact thermal imaging method. However, there were a number of issues that affected its deployment as a routine respiration rate monitoring system.



**(a)**



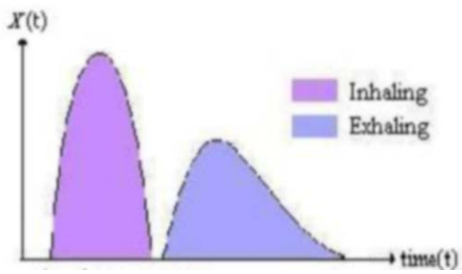

**(b)**

Fig. 2 (a) A thermal image, (b) respiration signal produced by thermal imaging

Small body movements did not affect the thermal image approach performance as an algorithm to deal with the effect had been devised. Very large body moments however, resulted in incorrect readings. The image processing of the extensive recorded images (i.e. 50 images per second) was not optimum, making real-time respiration rate monitoring not achieving. We are currently developing new techniques to deal with these limitations.

A video imaging technique for respiration rate monitoring using a webcam connected to a computer to detect chest movements was investigated [12]. The approach recorded images of chest movements at 10 frames per second. Chest movements were then detected by subtracting consecutive images. The pixel values in the resulting subtracted images were averaged to characterise each image by a single feature. This feature was plotted against time to produce a respiration signal. The respiration rate was then obtained from this signal by considering its average oscillations per-minutes. A typical set of images and a respiration signal cycle obtained using this approach is shown in Figures 3a and 3b. The first lobe of the respiration cycle in the signal is associated with air being inhaled, causing the chest to move forward and the second lobe is associated with air being exhaled, causing the chest to move to its original position. The figure shows that the inhalation interval is shorter than exhalation interval and there is a brief pause between the end of inhalation and the start of exhalation cycle. A limitation of the approach was that the subject was required to wear a patterned shirt with high contrast to allow chest movement detection to be possible by a single video camera.


**(a)**




**(b)**

Fig. 3 (a) A stack of video images during respiration recording, (b) one respiration cycle obtained from the video images.

The existing respiration monitoring systems have a number of limitations. These include the requirement for their sensing element to be attached to directly to the subject's body. These devices can only estimate the actual respiratory rate due to their short comings [13]. Respiratory rate is omitted from many children at an emergency department and the acute medical admissions unit of hospitals due to the unsettled nature of the patients for the devices to be attached on them [3]. In order to improve respiration rate monitoring, a portable device that performs respiration rate measurement in a noncontact and accurate manner is being developed. This senses the exhaled air at a distance and converts it to an electrical signal that in turn allows respiration rate to be determined.

In section 2 the structure of the air flow sensing respiration monitor is explained and its testing method is outlined. In section 3 the results are discussed. The conclusions are provided in section 4.

## 2. DEVICE STRUCTURE AND TESTING

The device evolved from an initial feasibility study [14]. A block diagram of the device is shown in Figure 4. A prototype heat chamber with two vents was made and a heater was enclosed. A circuit having a thermistor as the sensing element was also developed. The sensor was placed at the air outlet of the chamber and a funnel at its air inlet to direct expired air into the chamber. The components of the device are:

i. *Detachable Funnel:*To improve guiding expired air into the chamber'sinlet(14mm in diameter), a funnel was used. The funnel's radius was 75mm at the larger end and 13mm its smaller end. Its height was 65mm and it was made from a laminated paper.
ii. *Heat Chamber*: The chamber was made from a non-conducting material to minimize heat losses and to improve electrical safety since a heater was enclosed within it. It had an opening at both ends, where one was for air inlet and the other was for air outlet. The funnel was clipped to the inlet, whereas a thermistor was placed at its outlet to sense flow of heated expired air as it flowed out of the chamber. The chamber was carefully designed such as to achieve an effective heating and flow of the expired air. The heater operated at 5 volts and heated the expired air to about 50 $^{o}$C as it flowed through it.
iii. *Heat Sensing Element*: An Negative Temperature Coefficient (NTC) thermistor was used to detect the heated air flow. The resulting electrical signal was indicated the presence of expired air.
iv. *Electronic Circuitry*: The electronic circuitry filtered the signal from the thermistor using a 4$^{th}$ order Bessel filter. Bessel filter was chosen as it has a linear phase response. The filtered signal was amplified to provide full-scale input to the following analogue to digital converter. The digitised signal was read by a computer to display and process it in real time.
v. *Digital Signal Processing*: This part determined the respiration rate from the recorded respiration signal in both time and frequency domains.

The device was mounted on a fixed vertical position and held by a clamp such that the funnel was aligned with the nostril of a subject. A meter ruler was also clamped beside the chamber with a guide placed at the meter's origin which was also the position of the subject's nose. The distance between the funnel and the subject's face was changed between 15and 30cm to analyse the effect of distance on the device’s operation. Respiration rate was monitored while the subject was in a relax state. The recording was performed up to three minutes. The signal from the thermistor was sampled at 10 samples per second. The experiment set-up is shown in Figure 5.

The device was tested in a number of scenarios that included varying respiration rates and changing its distance from the face. The recordings were performed in a research laboratory (room temperature about 20 $^{o}$C) while the subject sat on a chair.

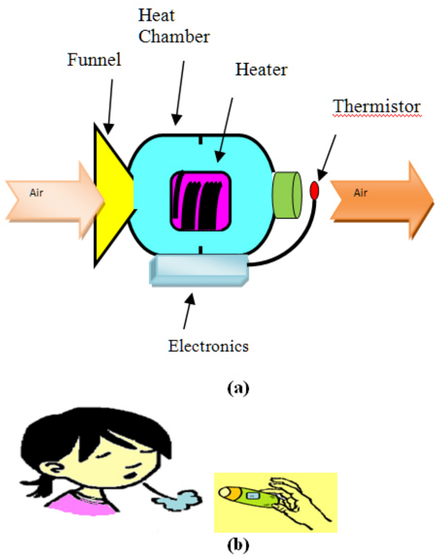



Fig. 4 (a) block diagram of the air-flow respiration rate monitoring device and (b) its application

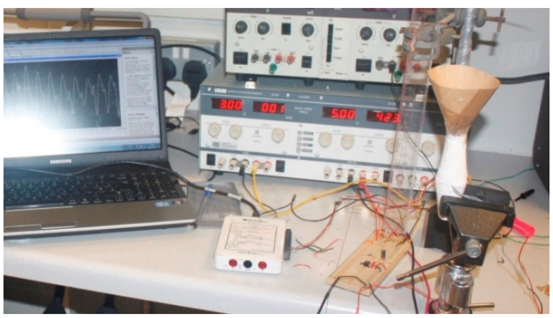


Fig. 5 Experimental Set-up

The results obtained when evaluating this device is presented in the next section.

## 3. RESULTS AND DISCUSSION

A typical respiration signal recorded when the device was held at a distance of 15 cm from the face is shown in Figure 5 (top). Its magnitude frequency spectrum is also shown. The peak of the magnitude frequency spectrum is at 0.25 Hz. This corresponds to one respiration every 4 seconds or 15 respiration cycles per minute. The respiration signal shown in Figure 6 is from a healthy adult subject.

In Figures 6 and 7, the respiration rates as seen from the magnitude of the frequency spectrums were about 17.6 per minute. In Figure 7(a), the point *p* shown by an arrow corresponds to a slight head movement away from the funnel. This indicates that as the nose had moved away from the funnel causing the amplitude of the respiration signal to be reduced. The respiration rate for the signal in Figure 8 is 55.6 cycles per minutes. The variations in the peak amplitude of the respiration signal was due to small head movements, both in sideways and up-down direction from the funnel.

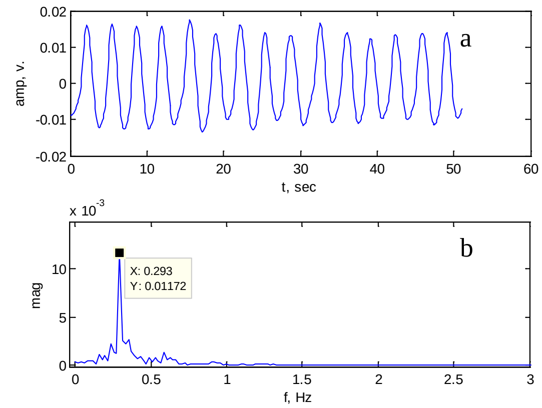



Fig. 6 (a) Respiration signal (at device distance of 15 cm) and (b) its magnitude frequency spectrum.

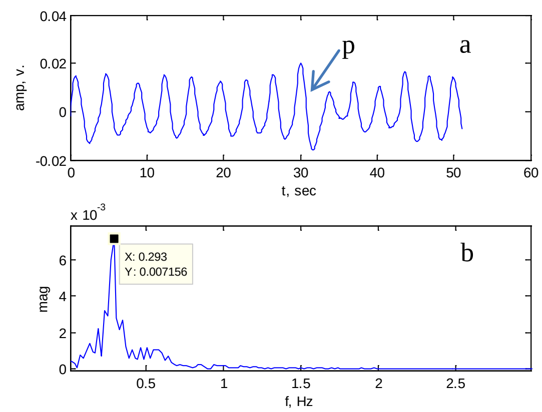



Fig. 7 (a) Respiration signal (at device distance of 15 cm) and (b) its magnitude frequency spectrum.

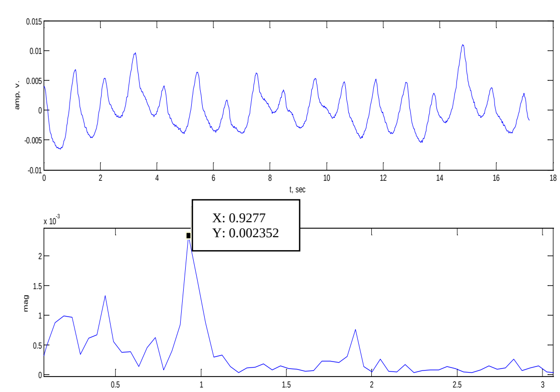



Fig. 8 (a) Respiration signal (at device distance of 15 cm) and (b) its magnitude frequency spectrum.

In Figure 9, the respiration rate was 10.5 cycles per minute. This signal points out the point where a deep breath is taken. The device could therefore indentify shallow from deep breathing patterns.

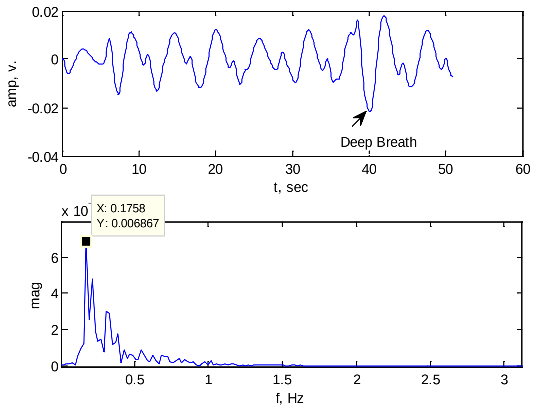



Fig. 9 (a) Respiration signal (at device distance of 15 cm) and (b) its magnitude frequency spectrum.

Figure 10 shows the signal when the device to facial distance was 20 cm. The respiration rate in this case is 14 cycles per minute. The signal amplitude is smaller as compared to that obtained for the 15 cm distance.

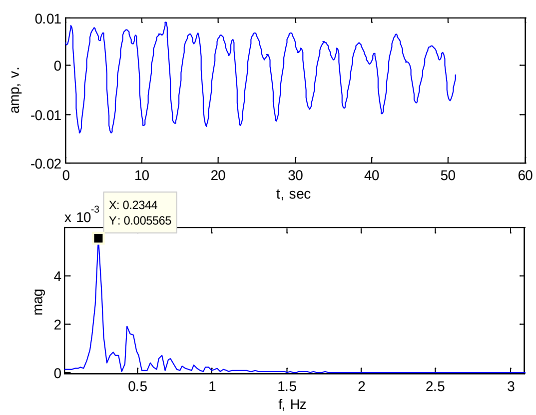



Fig. 10 (a) Respiration signal when the device distance from the face was 20cm and (b) its magnitude frequency spectrum.

The user interface of the device is such that the respiration signal is visible in real time on a PC interface. Figure 11 shows the respiration signal as measurements were taking place in real time. The device can be set to trigger an alarm when the respiration rate falls outside a predefined level.

The device is currently in its basic prototype form. We are currently working on optimising its performance so that it can be deployed in medical environments.

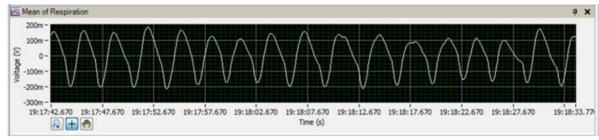



Fig. 11 Respiration signal during recording

## 4. CONCLUSION

A prototype expired air sensing respiration rate monitor device has been designed and its performance evaluated. The device provides a means of realising respiration rate monitoring in a noncontact manner. The results obtained indicate that the device was able to measure respiration effectively in a noncontact manner. It also detected deep breathing as well as head movements away from the device. Slight head movements in any direction affected the amplitude of the respiration signal, but not the respiration rate.

We are currently optimising the device's performance to make it deployable in medical environments.

## REFERENCES

[1] A. T. Johnson, C. G. Lausted and J. D. Bronzino, "Respiratory system," in *The Biomedical Engineering Handbook*, 2nd Edition ed., J. D. Bronzino, Ed. Boca Raton: CRC Press, 2000, pp. 7-1.

[2] A. Rakhimov, "Normal Respiratory Rate and Ideal Breathing," vol. 2012, 2012.

[3] F. Q. AL-Khalidi, R. Saatchi, D. Burke, H. Elphick and S. Tan, *Pediatr. Pulmonol.*, 46 (2011) 523-529.

[4] T. L. Lee-Chiong, Ed., *Sleep: A Comprehensive Handbook.* John Wiley and Sons Inc., 2006.

[5] K. Nepal, E. Biegeleisen and Taikang Ning, *Proceedings of the IEEE 28th Annual Northeast*, 2002, pp. 277-278.

[6] I. Konica Minolta Sensing , "Basic Understanding of the pulse oximeter," vol. 2012, 2006.

[7] J. James, L. Tiwari, P. Upadhyay, V. Sreenivas, V. Bhambhani and J. M. Puliyel, *BMC Pediatrics*, 6 (2006) 14-14, 2006.

[8] G. B. Moody, R. G. Mark, M. A. Bump, J. S. Weinstein, A. D. Berman, J. Mietus E. and A. L. Goldberger, *Computers in Cardiology*, 13 (1986) 507-510.

[9] M. Folke, F. Granstedt, B. Hök, H. Scheer, *J. Clin. Monit. Comput.*, 17 (2002) 97-103.

[10] Farah AL-Khalidi, Reza Saatchi, Heather Elphick and Derek Burke, *American Journal of Engineering and Applied Sciences*, 4 (2011) 586-597.

[11] I. Sens Comp, "MINI-A Ultrasonic Transducer," vol. 2012, 2004.

[12] K. S. Tan, R. Saatchi, H. Elphick and D. Burke, in *Communication Systems Networks and Digital Signal Processing (CSNDSP), 2010 7th International Symposium on*, 2010, pp. 770-774.

[13] R. Murthy, I. Pavlidis and P. Tsiamyrtzis, *Engineering in Medicine and Biology Society, 2004. IEMBS '04. 26th Annual International Conference of the IEEE*, 2004, pp. 1196-1199.

[14] S. McGrath, "Design and Development of a Non-Contact Respiration Monitoring Device," 2010.