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Head Phone Based Breath Rate Monitoring Using Pyroelectric Effect

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Abstract

This paper discusses a respiratory rate monitoring system that is based on piezofilm headphone. The respiratory airflow is sensed using a flexible piezofilm sensor attached to headphones. Inspiration and expiration phases vary in temperature during breathing. The Pyroelectricity of the piezoelectric film sensor detects this variation in temperature. Hence the subject's respiratory rate can be monitored with high accuracy by monitoring the variation in sensor output. Experimental results stipulates that the Pyroelectric response is faster than temperature-sensitive device thermistors. This type of head phone based proposed system makes it easy to be used for subjects' respiratory rate monitoring during sleep. The advantages of this respiratory rate monitoring system are rapid response time, low cost, and ease of implementation.

Keywords

Breathing rate, respiratory, airflow fluctuations, piezofilm, PVDF.

I. Introduction

Continuous monitoring of respiratory activity has importance applications in sleep studies, sports training, early detection of sudden infant death syndrome and patient monitoring pumotachographs are regularly used in the sleep research setting. They are not applicable for routine diagnostic sleep studies or other studies. The use of a pneumotachographs during sleep requires the patient to wear a nasal or face mask. The gold-standard device to measure flow is the pneumotachograph.[1] There, non-obtrusive sensors to detect breathing flow are especially suitable for those applications. During breathing,the air temperature is different between inspiration (room temperature) and expiration (approximately 37 °C) phases. Therefore, the respiratory rate can be derived from sensing the temperature variation. A new type of thermal sensor using polyvinylidene fluoride (PVDF) film has been reported to have a

faster response time than those of traditional thermal devices. [2] A PVDF film has both piezoelectric and pyroelectric properties. Caused by the pyroelectric property, the transducer's output signal is proportional to the temperature difference of the nasal airflow. The proposed system further uses this principle to measure the respiratory airflow. The respiratory airflow measurement is conducted by attaching a piezo film sensor to a headphone system placed close to the nasal airway.

II. Method And Materials

A. Piezofilm response to respiratory airflow

As a pyroelectric transducer is heated, the dipoles within the sensor exhibit random motion by thermal agitation. This phenomenon causes a reduction in the average polarization of thetransducer, generating a charge build up on the transducer's surfaces. The output current is proportional to the rate of

temperature change. The output voltage produced by thetransducer across the amplifier with an input resistance R_L is then given by

$$V_{OUT} = R_L Ap\left\{ \left.\frac{d\langle T\rangle}{dt}\right|_U - \left.\frac{d\langle T\rangle}{dt}\right|_D \right\} \quad (1)$$

Where*A* is the cross-sectional area of the transducer surface, *p* is the pyroelectric coefficient of the piezoelectric transducer,and *<T>* is the electrode's average temperature. Subscripts *U*and *D* denote the upstream and downstream electrodes,respectively. Equation 1 represents the principle of a,pyroelectric anemometer, [4] and can be used to measurerespiratory airflow for the proposed system.

B. Charge amplifier for piezoelectric transducer

To avoid the capacitance effect on sensitivity, a charge,amplifier circuit is applied to the piezoelectric charge readout.The charge amplifier eliminates the time constant effects of both the piezoelectric transducer and connecting cable. The amplifier is a current operated circuit with zero input impedance. This type of amplifier quickly absorbs charges produced by the transducer. This absorption action results in no charge left on the amplifier's electrodes and the transducer exhibiting no time constant.

Fig.1 shows the charge amplifier circuit. The signal charge pulses$Q_S$are generated by a piezoelectric transducer and then integrated to the feedback capacitanceC_f.The feedback capacitance$C_f$then produces voltage pulses $E_{OUT}(t)$.The output voltage pulses $E_{OUT}(t)$discharge with the time constantdetermined by $\tau = C_f * R_f$. If a piezoelectric transducer

provides a constant charge generation over a time interval of *t*= 0 to *t0*, then, using the Laplace transform, the output signal charge $Q_S$is given by the following equation [5]

$$Q_S(S) = Q_S\left(\frac{1}{S} - \frac{e^{-St_0}}{S}\right) \quad (2)$$

Similarly, the transmission coefficient *T(S)* is given by

$$(S) = \frac{1}{Cf}\cdot\frac{1}{S+1/\tau} \quad (3)$$

Therefore the output voltage $E_{OUT}(t)$is expressed using Eqn. 2 and 3 as follows:

$$E_{OUT}(S) = Q_S(S)\cdot T(S) = Q_S\left(\frac{1}{S} - \frac{e^{-St_0}}{S}\right)\cdot\frac{1}{C_f}\cdot\frac{1}{S+1/\tau}$$

$$= -\frac{Q_S}{C_f}\left(\frac{1}{S}\cdot\frac{1}{S+1/\tau} - \frac{e^{-St_0}}{S}\cdot\frac{1}{S+1/\tau}\right) \quad (4)$$

As a result, the output voltage$E_{OUT}(t)$ is given by

$$E_{OUT}(t) = -\frac{Q_S}{C_f}\cdot\frac{1-e^{-1/t}}{t_0/\tau} \qquad 0 \le t \le t_0$$

$$= -\frac{Q_S}{C_S}\cdot\frac{(e^{t_0/\tau}-1)}{t_0/\tau}e^{-t/\tau} \qquad t_0 \le t \quad (5)$$

Because $t_0 << \tau$, Eq. 5 can be simplified as follows:

$$E_{OUT}(t) = -\frac{Q_S}{C_f}e^{-t/\tau} \quad (6)$$

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As can be seen from Eq. 6, the charge produced by the piezoelectric transducer Q_S is converted into voltage pulses that are damped by time constant τ with amplitude E_{OUT} as follows:

$$V_{OUT} = -\frac{Q_s}{C_f} \qquad (7)$$

From Eq. 7 it is evident that the output voltage E_{OUT} is directly proportional to the electric charge yielded by the transducer.

C. Experiment Setup

To evaluate the performance of the proposed system, we captured the response waveform of the system with respiratory airflow from the subjects' nose. Figure 2 demonstrates the proposed system setup. APVDF type piezo film is attached on a headphone. The output of the pyroelectric transducer is first amplified by a pre-amplifier circuit. The output is then sent to the filter circuit. Finally, the signal is amplified by a gain adjustable amplifier.

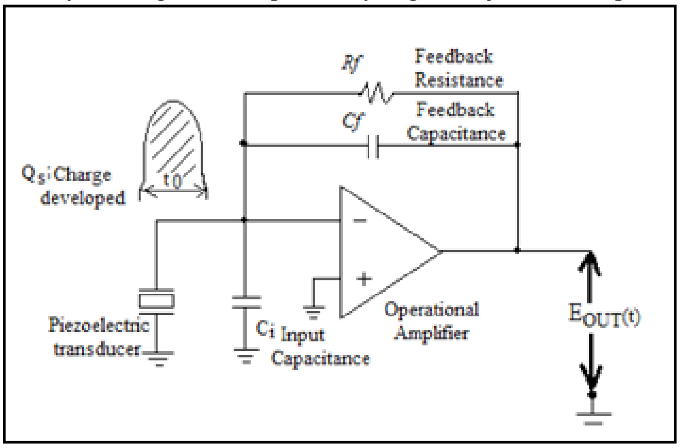

Fig .1 : Piezoelectric transducer's charge amplifier circuit.

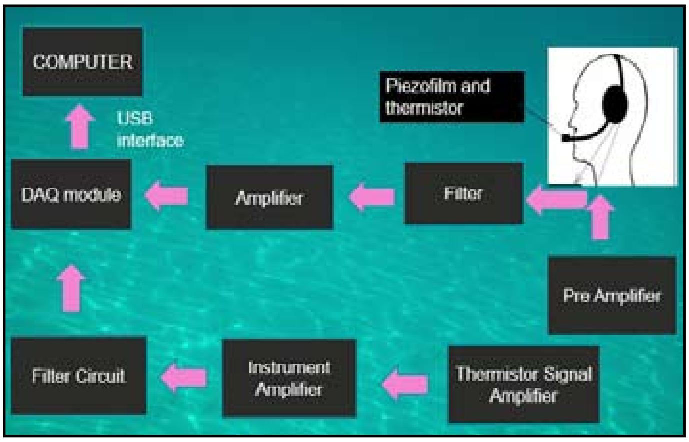

Fig. 2. Experiment setup of the proposed system

To evaluate the proposed system response time, the experiment compares the proposed system's performance to a temperature sensitive transducer. A negative-temperature coefficient (NTC) type thermistor (NTSE0103JZ238, Thinking Electronic, Taiwan) is installed alongside the piezoelectric transducer. The thermistor has a resistance spec of 10 kΩ at 25°C and 3.496 kΩ at 50 °C. It is capable of temperature measurements ranging from 0 °C to 85 °C. The signal generated from thermistor is first amplified by a thermistor amplifier (INA330, Texas Instrument,USA). The output of the first stage amplifier is then further processed by an instrument amplifier (INA114, Texas Instrument, USA) and is sent to a filter circuit. The filter is composed of a first-order high-pass filter (cut-off frequency 0.07 Hz), a second-order Butterworth low-pass filter (cut-off frequency 100 Hz), and a 60 Hz bandrejectSallen-Key filter.

A belt system is used as the gold standard for benchmarking the respiration rate measurement of our system. A piezo-respiratory belt transducer wrapped around the subject's diaphragm measures the thoracic circumference during expiration and non-expiratory phase. The output of the belt is amplified by an instrument amplifier (INA114, Texas Instrument, USA), and filtered by a low-pass filter (cut-off frequency 100 Hz) and a 60 Hz band-reject filter. Both the outputs of the belt system and the proposed system are acquired by an interface module (UIM100C, Biopac System,USA) and then sent to a personal computer via USB port foranalysis and comparison.

D. Experimental Breathing Types

A human breathing cycle consists of inspiration, expiration,and post-expiratory pause phases. During quiet breathing, inspiration is initiated by chest cavity pressure caused by the diaphragm's negative contraction. Expiration is a passive process where the airflow is produced due to the lungs' elastic recoil property. The post-expiratory pause is caused when the pressure inside and outside the lungs is equalized. A breathing cycle is defined as the time interval between the beginning of inspiration and the end of the post-expiratory pause phases. Tocompute the respiratory rate, the inspiration, expiration, and post-expiratory phases are timed. The three phases are summed together to derive the breathing cycle time T_C by

$$T_C = T_{IN} + T_{EX} + T_P \qquad (8)$$

where T_C, T_{IN}, T_{EX} and T_P represent breathing cycle time, inspiration duration, expiration duration, and post-expiratory pause duration. Additionally, the respiratory rate is computed in cycles/min by the following equation

$$\text{Rate} = 60\ T_C \text{ cycles /min .} \qquad (9)$$

Normally after physical exertion, the post-expiratory duration is reduced considerably. In some cases this phase mayeven be reduced to zero. As a result of breathing differentialsresulting from physical activity, our experiments examine two different breathing types. The first type (Type I) examines breathing after the subject has stepped on and off a stepping stool 30 times in 30 s. The Type I breathing mode consists of only inspiration and expiration phases. [6] The second breathing mode (Type II) examines quiet breathing while the subject is resting in a chair. The Type II breathing mode consists of inspiration, expiration, and pause phases. Both Type I and Type II breathing modes are evaluated in the pneumotach and thermistor experiments.Each experiment system records 25 Type I and Type II breathing cycles for each subject. To reduce the effects caused by installing the mask on the subject's face, the data of the first 5 cycles is disregarded. Therefore, the data from each experiment contains 20 breathing cycles for analysis.

III. Literature Survey

1. Noninvasive monitoring of respiratory mechanics during sleep

Farré R, Montserrat JM, Navajas D

METHODOLOGY- Changes in patient ventilation are assessed by recording flow or volume signals by means of pneumotachographs, thermistors or thermocouples, nasal prongs or thoraco-abdominal bands. Common tools to noninvasively assess breathing efforts are the thoraco-abdominal bands and the pulse transit time

technique.
APPLICATION- They allow a semi quantitative assessment of the physiological variables. These techniques are focused to monitor different aspects of the sleep-related breathing disturbances (airflow, effort or obstruction).This also provide redundancy and consistency in the detection of respiratory sleep events, both during hospital polysomnography and in the home-monitoring setting.
LIMITATION- The technical and practical limitations of each technique, combining different tools improves the reliability and robustness of patient assessment during sleep.

2. AURA: a new respiratory monitor and apnea alarm for spontaneously breathing patients

Cyna AM1, Kulkarni V, Tunstall ME, Hutchison JM, Mallard JR.
METHODOLOGY- Respiratory Alarm allows monitoring by utilizing the pyroelectric property of polarized polyvinylidine fluoride sensors. A quartz crystal oscillator generates pulses that allow measurement of inter expiratory time and ventilatory frequency. The system incorporates LED digital displays, a bargraph and audiovisual alarms.
APPLICATION- Detect temperature changes that occur during breathing into an oxygen delivery face mask
LIMITATION- Excessive movements involving displacement of the mask produce signal artefacts and false positive alarms.It was felt unlikely that these apnoeic episodes would have been noticed if the alarm had not been activated. Setting the gain control at too low a value led to false positive alarms.

3. Pyroelectric anemometry: Theory of operation

H. Y. Hsieh, H. H. Bau, and J. N. Zemel.
METHODOLOGY-The pyroelectric anemometer (PA), as a function of the laminar flow rate. The critical Reynolds number, , is a function of the gas's thermophysical properties, the thermal excitation frequency, and the spacing of the measuring electrodes relative to the central heater strip.Convective heat loss at moderate flow rates and low flow rate.
APPLICATION- This successfully predicts the dependence of the PA's response on geometry, operating frequency, and thermophysical properties of the gas. The model is useful for obtaining insights into the PA's operation and as a design tool for future PAs
LIMITATION- Significant differences are also observed for the six-electrode PA's second and third electrode pairs. These deviations are expected because of the relatively poor agreement between the theoretical and experimental values of A,, and BP While Re,, in general, is about half of the experimental values, the general agreement confirms that ReE is proportional to frequency.

4. Heat transfer model for the pyroelectric anemometer

P. Hesketh, B. Gebhart, and J. N. Zemel
METHODOLOGY-A resistor□capacitor analog heat transfer model is used to calculate thermal pyroelectric anemometer (PA) response. The pyroelectric substrate is thin, the analysis is simplified to one dimension and excludes edge effects, heat losses due to the support, and thermal radiation. Further, no temperature difference is assumed at the interface between the gas and substrate, i.e., no temperature jump.
APPLICATION- A thermal resistor-capacitor analog model is developed to analyze the data.
LIMITATION- Regardless of the causes of the discrepancies between the magnitude of the calculated results and data, the functional agreement offers the opportunity to theoretically explore the behavior and response of the PA geometry.

5. Noncontact measurement of breathing function

Murthy, R. and I. Pavlidis
METHODOLOGY- Photoplethysmography (PPG) is a variant method of the ECG, developed to measure blood volume changes in living tissues by absorption or scattering of near-infrared radiation. This modality consists of an infrared light-emitting diode (LED) and a photodiode that can be clamped to the ear lobes, thumbs, or toes.
APPLICATION- It is advantageous because it is portable, compact, and needs very little maintenance. The measurement of blood volume changes by PPG depends on stronger absorption of near-infrared light by blood when compared to other superficial tissues. It can be used to predict various life threatening disorders like sudden infant death syndrome and heart attacks. It is also used in sleep studies to detect sleep apnea.
LIMITATION-Measurements by these methods are corrupted either by movement artifacts or by their dependence on other physiological variables, like heart rate.

IV. Result

Figure 3 shows the respiratory waveforms derived from the belt system and the proposed system. The inspiration, expiration, and post-expiration phases are marked in different colors. It can be seen that the proposed system can be used to accurately measure a subject's respiratory phases. In addition, experimental results indicate that some airflow fluctuations appear only in the proposed system response waveform. The belt system is unable to detect these airflow fluctuations. Figure 4 shows the result of respiratory rate calculation. When using the belt as a benchmark, the average period time error for the proposed system is 67.2 ms. The 67.2 ms system error translates to an average 0.45 (cycles/min) and 1.12% average respiratory rate error. Therefore, when compared to the belt, the accuracy of the proposed respiratory measurement system is approximately 98.88%.

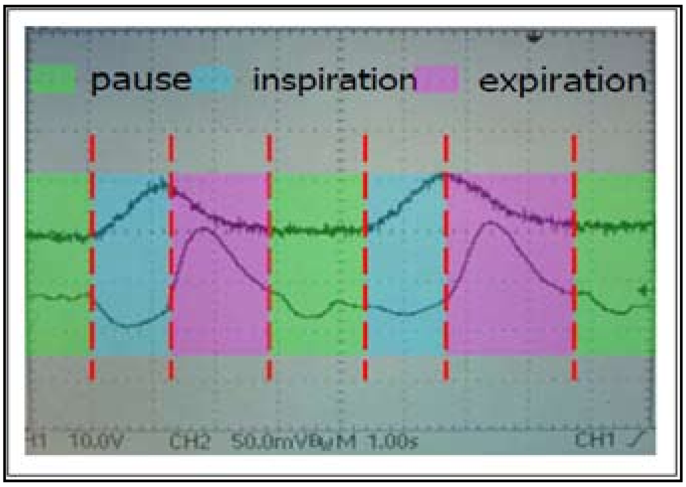

Fig .3 : Respiratory waveform derived from boththe belt and the proposed system.

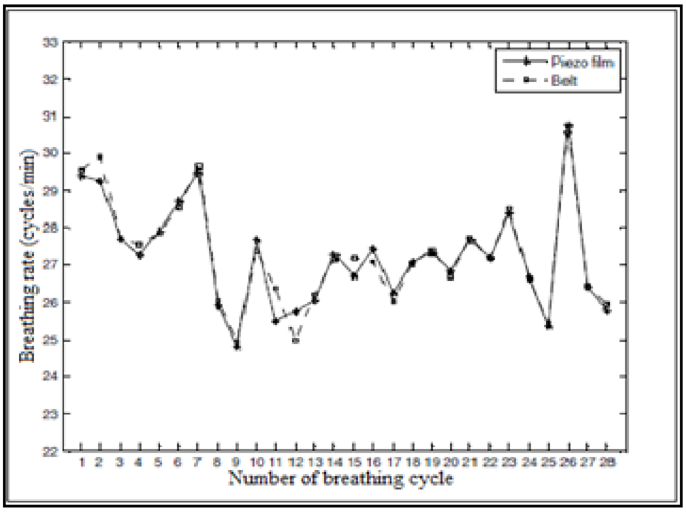

Fig .4 : Result of the respiratory rate measurement.

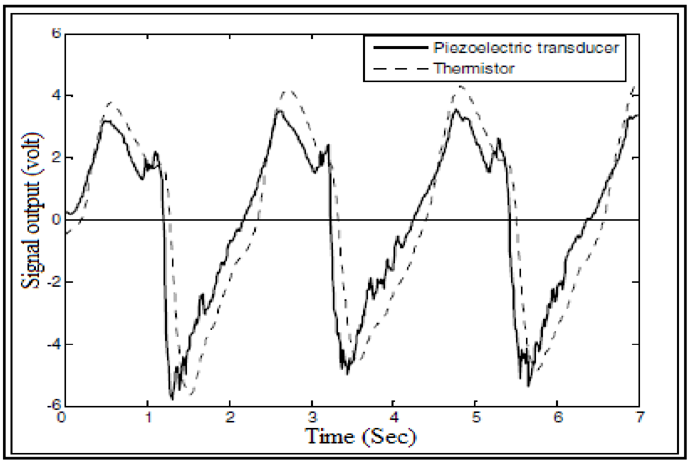

Fig .5 : Type I breathing output waveform of onesubject from the proposed system and the thermistor

Figure.5 shows three of Subject one's twenty-eight Type I breathing outputs from the proposed system and the thermistor. The phases which have positive amplitude output are inspiration phases and phases with negative amplitude output are expiration phases. Figure 5 shows that the thermistor output waveform is smoother than the proposed system output waveform. Therefore some airflow fluctuations appear only in the piezo film transducer response waveform. These airflow fluctuations were not detected by the thermistor. Although the thermistor is installed alongside the piezo film transducer, the thermistor response waveform slightly lags behind the piezo film transducer.

V. Discussion

The proposed transducer system can accurately measure respiratory rate. When using a belt as a benchmark, the accuracy of the proposed respiratory measurement system is approximately 98.88%. Furthermore, experimental results indicate that some airflow fluctuations appear only in the proposed system response waveform. In addition, the propose piezo film system has faster response time when compared with a thermistor. The advantages of this respiratory rate monitoring system are rapid response time, low cost, and ease of implementation.

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