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# "Are you breathing?" – Design, build and testing of a low-cost, portable respiratory rate monitor

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**Abstract:** A patient's respiratory rate is one of the critical vital signs that is a determinant of patient well-being. However, it is all too often neglected or misreported by health care professionals. This study presents the design, build and testing of a low-cost, portable monitor to facilitate accurate reporting of respiratory rate. The monitor comprised a thermistor-based transducer to capture the breath cycle of patients based on the temperature differential created across the thermistor. The signal was conditioned and processed such that the signal could be analysed to identify the peaks and ultimately determine the respiratory rate. For a total cost at the time of development of less than €40, the integrated system demonstrated a modest average error of 5.6 % across a range of different ambient temperatures, rate and depth of breathing, and orifice of breathing. This is comparable with other commercial and custom devices. The presented monitor may be of interest for use in an emergency room or clinical setting, especially in severely resource-constrained countries.

**Keywords:** Vital sign, minimally-invasive monitoring, peak detection, resource-constrained solution

## 1 Introduction

Monitoring a patient's vital signs, including the respiratory rate, is indispensable to health care professionals since they are understood to be determinants of patient well-being [1, 2]. Many of the available methods of respiratory rate monitoring are expensive, lack portability, and can cause patient discomfort [3]. This warrants the development of a compact, portable, affordable, and minimally-invasive means of rapidly and reliably determining a patient's respiratory rate.

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There are many methods of determining respiratory rate, ranging from non-contact approaches such as video analysis of patients, to the more common contact methods based on various physiological variables [3–7]. These methods rely on measuring one or many physiological variables which are associated with respiration, including blood $O_2$ levels, breath humidity and temperature, and breath $CO_2$ concentrations [8, 9]. Regardless of the measured physiological variable, a respiratory rate monitor must capture the signal of the associated peaks and troughs of inhalation and exhalation during the breath cycle, ultimately determining the respiratory rate from the recorded signal.

The gold standard of respiratory rate monitoring is capnography, which has an error of 2.6 % relative to manual measurements [10]. Some commercially available devices have errors between 4.7 % and 8.3 % [10, 11]. Other custom devices have errors between 5.6 % and 11.2 % [3, 4, 12]

In this study we present the design of a low-cost portable respiratory rate monitor. Following construction, the monitor is tested to verify the accuracy of the respiratory rate measurements recorded.

## 2 System design

An overview of the respiratory rate monitor is shown in Figure 1, the associated circuit diagram in Figure 2, and the physical device in Figure 3. There were broadly three key elements to the system, namely the signal acquisition and conditioning, the digital processing and local display, and the remote logging and data storage. The specific components chosen for the final design were based on a trade-off of performance and cost.

### 2.1 Electronic components

The respiratory rate monitor comprised a G10K3976 radial glass negative temperature coefficient thermistor (Measurement Specialties Inc., Shrewsbury, MA, USA) that was incorporated into a deflection bridge. The thermistor was attached directly to a nebuliser mask, such that it could be in the immediate pathway of a patient's breath, then connected to the circuit through a shielded twisted-pair wire. The output sig-

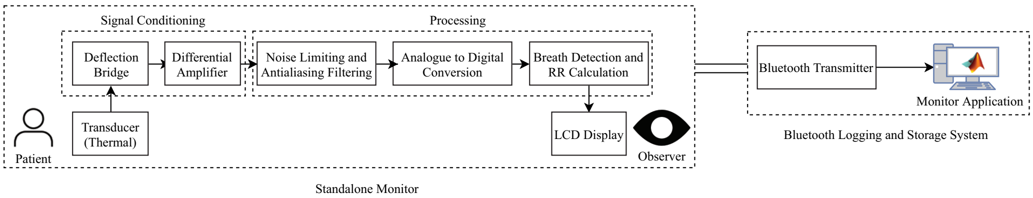


**Fig. 1:** System diagram of the respiratory rate monitor, from the patient breathing into the standalone device to the display for the observer. The remote data logging and storage is an optional module that can be selectively included.

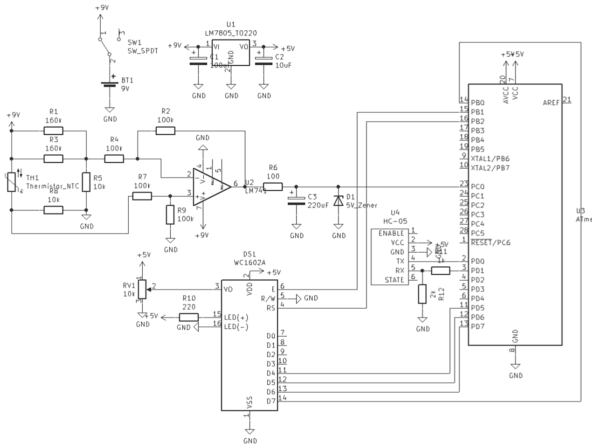

**Fig. 2:** Circuit diagram of the standalone respiratory rate monitor, with the optional bluetooth module connections shown.

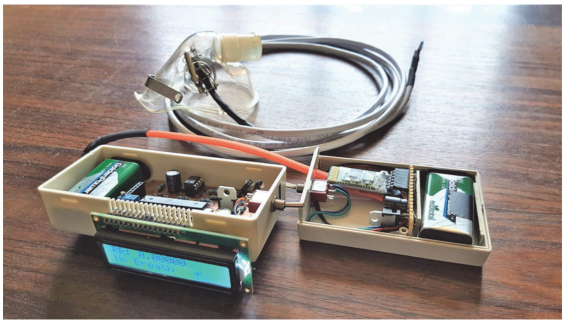

**Fig. 3:** Photograph of the respiratory rate monitor (lower left) and nebuliser mask housing the thermistor (top) alongside the optional bluetooth transmitter circuit (lower right).

nal from the deflection bridge was passed through a unity-gain differential amplifier built using an AD820 operational amplifier (Analog Devices, Inc., Norwood, MA, USA). To attenuate high frequency noise and reduce error due to aliasing effects, a first order low-pass passive filter with a cut-off frequency of 7-Hz was implemented. Digitisation and further processing was performed on an ATmega328P microcontroller (Atmel Corporation, San Jose, CA, USA). The measured respiratory rate was shown on a generic 16×2 liquid crystal display. An optional HC-05 bluetooth module (SUNLEPHANT, Guangdong, China) could be selectively incorporated into the system in a plug-and-play fashion for remote monitoring and data storage for offline analysis. The electronic components were powered from the ubiquitous PP3 9-V battery, with a LM7805 5-V regulator (STMicroelectronics, Geneva, Switzerland) providing the regulated lower voltage supply. The optional bluetooth module was powered from a separate supply.

## 2.2 Digital processing

The analogue signal was converted to a digital signal using the built-in ADC of the microprocessor at a sampling frequency of 120 Hz. The bandwidth of the signal was then further reduced to 3 Hz using a 30$^{\text{th}}$ order finite impulse response low-pass filter. The intention of this filtering was to remove possible noise introduced during the conversion process. To determine the respiratory rate from the measured signal, a peak detection algorithm was developed. Rather than classical peak detection, the implemented algorithm effectively searched for a change in polarity of the gradient to determine the shift from expiration to inspiration. To ascertain the gradient, ordinary least squares regression was used, together with a sliding window across the measured sample points. The gradient change was determined about a hysteresis band rather than a zero-crossing point for additional noise immunity. Finally the respiratory rate was calculated by taking the weighted average of the intervals between the last five detected breaths. Once the respiratory rate had been calculated, it was displayed on the standalone device. If the optional bluetooth module was connected, the data was also transmitted to a remote logging station for capturing and further processing.

# 3 Testing methodology

To verify the capabilities of the respiratory rate monitor the testing was performed in three stages. The analogue testing characterised the system response of the transducer and signal condition circuit. The thermistor was placed in an impermeable bag and submerged in a tank of water, initially at 40 °C, and the output voltage of the analogue system measured as

**Tab. 1:** Test criteria of the integrated respiratory rate monitor. *Temp.* is the ambient environment temperature in °C, *Rate* is the rate of breathing in breaths per minute [13, 14], *Depth* is the depth of the breathes taken, and *Orifice* is where the breath originates.

| Condition | Criteria | | |
|---|---|---|---|
| *Temp.* | Cold (<10) | Room (23–25) | Hot (>28) |
| *Rate* | Slow (<16) | Normal (16–20) | Fast (>20) |
| *Depth* | Shallow | Normal | Deep |
| *Orifice* | Nasal | Oral | |

the temperature was decreased to a minimum of 0 °C. The digital testing characterised the system response of the respiratory rate detection algorithm. A sinusoidal output from a signal generator was used to simulate a patient’s breath for a range of frequencies and amplitudes. The integrated testing demonstrated the full functionality of the respiratory rate monitor. Five test subjects participated in a total of 131 experiments of between 30–60 s to validate the performance across a range of possible input conditions summarised in Table 1. The nebuliser mask was gently held over the subject’s nose and mouth and their breathing pattern and reported respiratory rate was recorded, then transferred to a remote logging station using the monitor and optional bluetooth module. The MATLAB® (The MathWorks, Inc., Natick, MA, USA) *findpeaks* function served as a point of comparison to a classic peak detection algorithm. Additionally, for each experiment the peaks were also manually identified to create the ground-truth data to determine the accuracy of the integrated system. The data was recorded and processed using MATLAB®.

# 4 Results and discussion

The relationship between temperature and output voltage is shown in Figure 4. The sensitivity of the analogue components of the respiratory rate monitor is 100 mV / °C. The output voltage relative to temperature demonstrates a low level of non-linearity of approximately 6 % across the operating range. Thus the temperature could readily be inferred from the voltage signal if desired. When operating over an extended period of time, as the voltage of the battery supplying the circuit degraded to 8 V, the average output voltage from the analogue components decreased by an average of 11 %. However, this did not significantly affect the detection of individual breaths since the algorithm was not based on a preset threshold value.

A representative example of the data recorded from the breath detection algorithm and reported breaths is shown in Figure 5. There is an inherent delay between the change of gradient of the recorded signal and the reporting of a breath.

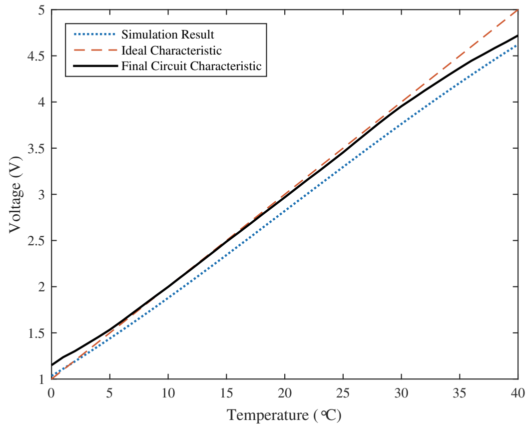


**Fig. 4:** Comparison of the characteristic system response of the analogue component of the device (—) to the ideal response (– –) and the simulated response (····).

This is primarily due to the detection method employed, but there were additional delays as a result of the high order digital filter introduced after the ADC. At the different ambient temperatures, breath rates and depths, and orifices the average delay in the algorithm was 0.22±0.09 s to reporting the breath from when it happened. This delay does not affect the accuracy of the monitor since an unrealistic respiratory rate of 120 breaths per minute is still slower.

Using the calculated respiratory rate from the manually marked peaks as reference, it was found that there was an average error of 5.6 % across all experiments from the device. In contrast to this, when compared with the *findpeaks* function there was an average error of 21.1 %. The relative errors for the various test categories is summarised in Figure 6. This relative error is significantly higher when compared with the *findpeaks* function due to the number of false peaks reported from the function. The relative error of the device was lower at the cold and room temperatures. At higher ambient temperatures, the temperature differential between the inhaled and exhaled air across the thermistor is lower. This ultimately leads to false or no detections due to insufficient changes in gradient which limits the environment that the device could be used in. The performance of the device was consistent for the different depth of breaths, and was found to accurately detect respiratory rates up to 120 breaths per minute reliably. It is hypothesised that the accuracy of the device is better for oral breathing due to a greater volume of air being drawn over the thermistor in a shorter space of time.

It should be noted that at the time of development, the total cost of the components for this system was less than €40. This device is inexpensive and simple to implement which can be used in an emergency room or clinical setting to monitor the patient’s respiratory rate.

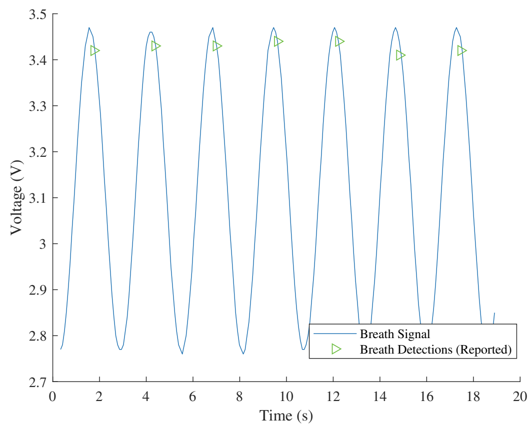


**Fig. 5:** The measured signal (—) of a simulated respiratory rate of 23 breaths per minute. The reported detection of the breath is indicated by green triangles.

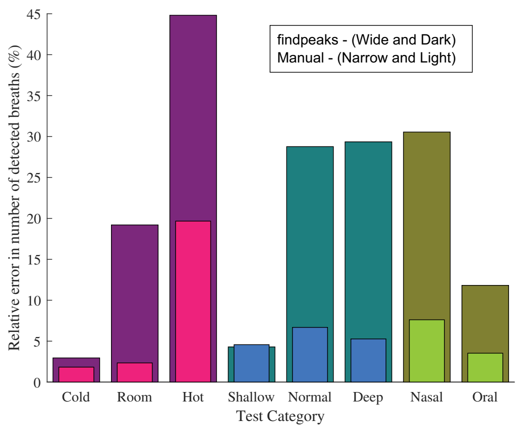


**Fig. 6:** Relative error in breath detection grouped by temperature, depth of breathing, and breathing orifice. The different depths of breath have been combined for each of the other test categories. The wider/darker bars represent the relative error compared to the *findpeaks* function, and the narrower/lighter bars represent the comparison to the manually identified true peaks.

# 5 Conclusion

In this study a low-cost portable respiratory rate monitor was produced and tested. The integrated system had an average error of 5.6 %, which is comparable with existing commercial devices, indicating that it is a reliable method of measuring a patient’s respiratory rate for a low cost. This device may be of interest for use in an emergency room or clinical setting, especially in severely resource-constrained countries.

### Author Statement

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