**Low-Cost, Thermistor Based Respiration Monitor**

Hana Qudsi, Maneesh Gupta

*Abstract*—Respiration rate is a standard physiological measurement taken for monitoring a patient. Existing respiration monitors used in medical centers are too expensive to be used in low-resource environments. We have developed a low-cost respiratory monitor to address this issue. This device calculates a patient's breathing rate by detecting changes in temperature through a thermistor when the patient breathes in a mask. Features of the device include an alarm through a piezoelectric speaker which sounds when the patient stops breathing and a low-battery indicator signal for when the battery powering the device dips below a threshold voltage. The design and implementation involved several different hardware components, as well as software functions. Our initial prototype of our project accomplished our main objectives, and experimental measurements indicate that the device measures a person’s respiration rate with relative accuracy.

# Introduction

With our project we wanted to address a real-world problem. Many biomedical devices, such as commercially available respiratory monitors, are expensive and require a stable power-supply to operate. We identified a critical need for respiration monitors that were inexpensive and easy to use, and thus implemented a solution that is adaptable to different environments.

When examining different methods of measuring respiration, we came across several commercial respiratory monitors using different types of sensors. Some use a chest force sensor to detect force produced by chest movement, while impedance pneumograph monitors use skin electrodes to calculate transthoracic impedance. The last common method is pulse oximetry which measures O2 saturation in blood. We aimed for our thermistor-based monitor to be comparatively cheaper, easier to use, as well as reliable.

Hana Qudsi is with Cornell University, College of Engineering, Ithaca, NY 14853 USA (e-mail: htq2@ cornell.edu).

Maneesh Gupta is with Cornell University, College of Engineering, Ithaca, NY 14853 USA (e-mail mg573@cornell.edu)

# Project OverView

To use our device, a patient wears the nebulizer mask which has an embedded thermistor, a temperature sensitive resistor. As the patient breaths into the mask, the temperature across the thermistor oscillates in parallel with the patient’s respiration rate. Thus, a measurement of the voltage across the thermistor voltage divider can be used as an indirect measurement of the respiration rate.

A patient’s respiration rate is measured and displayed in breaths per minute (bpm) on a small LCD screen. This allows whoever is monitoring the patient to see if the patient is breathing too fast or too slowly. Our device generates two alarms by a piezoelectric speaker; to indicate when a patient is not breathing and to signal that the device is running on low battery. This design allows for reliable use in environments where battery power may be necessary.

# Hardware design

## Thermistor Amplification

During experimentation we found that the change in voltage from a person breathing on the thermistor was on the order of 10 mV. In order to better detect a “breath” with the microcontroller, we amplified this change using an operational amplifier and high pass filter. We choose the time constant for our high pass filter to be 22 seconds so that breathing with a period less than 22 seconds will be amplified by our operational amplifier. After amplification, the change in breathing was in the 0-5 V range.

## Analog Comparator

The respiration rate is measured by using an analog comparator (AC) to compare the amplified voltage across the thermistor to a reference voltage. Since the respiratory signal is approximately equivalent to a sine wave with a DC offset, we choose the reference voltage to be at the DC offset. However, patients of different ages will have different steady state breathing temperatures, and thus DC offsets. To automatically determine the reference voltage for a patient, we use a low pass filter with a time constant of 22 seconds. Since the respiratory signal gets filtered, the output of this filter is the average value of the incoming signal. As the incoming signal is equivalent to a sine wave with a DC offset, the average value of the signal is the DC.

## Battery Monitor

To measure the voltage of the 9-V battery we use a voltage divider along with a low pass filter. The voltage divider is used to convert the voltage range from 0-9 V to 0-5 V so that the microcontroller pins will not be damaged. The purpose of the low pass filter is to eliminate any high frequency signals that might occur. The time constant of the filter is 1 µs and the resistance values for the voltage divider were chosen so that the output was one quarter of the original voltage.

# SOFTWARE DESIGN

The C programming language was used to program the AtMega 1284p microcontroller, using AVR Studio 4.15 and WINAVR GCC C compiler.

In the main execution loop, the device goes through a series of conditions to see if a specific task needs to be called. The thermistor and sampling tasks are both timed, so if either timer count equals 0, the counter is reset and the task is called. The thermistor sampling task is called every second, while the battery sampling task is called every minute. The measure task is called when the “measure flag” indicates that a measurement is ready to be taken. Lastly, the display task activates when the user holds down the display button.

To detect breathing, the absolute value of the difference between two consecutive samples from the ADC is stored. If this difference is less than or equal to 8, then this is considered a "no breath". If the difference is above the tolerance, then the "no breath" counter variable is reset to 0 and the pulse-width modulated (PWM) signal is set to be turned off. The tolerance for the difference between breaths was found through experimentation.

A counter is used to keep track of the number of "no breaths" detected because depending on how a person is breathing, consecutive samples could fall into the tolerance range. After 10 “no breaths”, the alarm sounds.

Similarly for determining a low battery, if the battery sample is found to be less than or equal to the tolerance, a PWM signal is generated, left on for about 1 second, and then is turned off. The PWM channel which produces a tone for a low-battery warning has a larger pre-scalar than the PWM channel to produce the alarm for when a patient is not breathing, making the low-battery tone lower in pitch.

The respiration rate is calculated by measuring the time difference between two samples when the voltage across the thermistor crosses the reference voltage. This is done in an interrupt service routine (ISR), which measures the difference in the number of clock cycles between two reference voltage crossings. The “measure flag” is then set, activating the measure task. In the measure task, an average difference in clock cycles is calculated by doing a weighted average between the newly measured difference and the previous average difference. The weights are 0.75 of the previous average and 0.25 of the new difference. This weighted average is taken so that an average respiration rate can be calculated. This average difference in clock cycles is then converted to breaths per minute. The display task, which displays this calculated value was borrowed from a preexisting library of functions written for this LCD.

# testing and simulations

Characterization of the device began by measuring the voltage across the thermistor in the voltage divider. The voltage across the thermistor was measured while the experimenter breathed on the thermistor in the mask, displaying a sinusoidal waveform that matched the experimenter’s breathing. To test the sensitivity of the thermistor, the experimenter varied his breathing, demonstrating the near instantaneous response of the displayed voltage.

Next, this signal was sent to the AC to test how accurately the respiration rate was measured. This was done by connecting the thermistor to the oscilloscope to measure the actual respiration rate. While this was being done, the AC was used to measure the respiration rate and the two numbers were compared. The accuracy of this device was found to be within 5% of the actual respiration rate.

The next piece of the device that needed to be tested was the alarm system. The first alarm is activated when the patient is not breathing. This was simulated by having the experimenter breath into the mask for several minutes before quickly removing the mask from his face. After numerous repetitions it was found that the alarm would activate after 5 seconds.

The second alarm activates when the battery voltage is lower than 7.5 volts. To simulate this situation a power generator with a variable voltage was used as the power source for the device. The voltage was then varied above 7.5, at 7.5 and below 7.5. At 7.5 volts and below the alarm sounded, but did not sound above 7.5 volts.

# Summary

This device can accurately measure a respiration rate, output an alarm if the patient is not breathing, measure a battery’s voltage, generate a low-battery indicator, and turn on the LCD only when a button is being held down. The piezoelectric speaker will sound an alarm within 5 seconds of when the patient stops breathing. This is an extremely important accomplishment because it is imperative to know as soon as possible when a patient is not breathing.

We hope to make some improvements in the next iteration of our device. There is still some lag between when the device is turned on and when the device begins measuring respiratory rate, which we would like to see minimized. We also are looking to improve the accuracy of the measured respiration rate. This can be done by either adjusting the weights of the current algorithm, or by using a different averaging method. The current method is not optimal because of the delay in converging to a final value and its insensitivity to change. Satisfying both of these requirements is difficult to accomplish since satisfying one generally means compromising the other. The final cost for this device is approximately $39.00, however this can be further reduced through optimization of the components and mass production techniques.

# acknowledgment

We would like to thank Professor Bruce Land, Senior Lecturer at Cornell University, and his course staff for their encouragement, inspiration and guidance.