Obstructive Sleep Apnea Device

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***Abstract* — Obstructive sleep apnea blocks the upper airways, causing a person to stop breathing in their sleep, making for restless sleep. One way to track these changes is by tracking their breathing frequency. This paper designs a circuit aimed at tracking breathing patterns and alerting users while in their sleep to avoid the choking sensation associated with obstructive sleep apnea. This breathing data is primarily found using a sensitive thermistor to transduce the pattern of inhaling and exhaling, then processing that signal with 555 timers and logic gate, and finally leading to an alarm that goes off when the time of non-breathing exceeds 90 seconds. Virtual simulations of the circuit were conducted, attempting to confirm the validity of our design. Alternative and additional components to explore in the future to further improve the design were also included.**

I. Introduction

A. Motivation

Obstructive sleep apnea (OSA) is a disorder that affects a person's ability to breathe while asleep due to blocking of upper airways. People with OSA report choking or gasping while sleeping, restless sleep, and difficulty staying asleep. This disorder affects 2-9% of Americans, often going undiagnosed [1]. Additionally, the disease has been shown to be a risk factor for heart diseases, even as far as sudden cardiac death [2].

Current solutions for OSA include surgery, oral appliances, or more commonly continuous positive airway pressure (CPAP) machines. While CPAP machines are shown to be effective, about 40-70% of patients with sleep apnea will stop using CPAP machines [3]. Despite the disorder persisting, they often stop using the machine due to discomfort with the CPAP mask, nasal irritation, or belief that they no longer need to treat their OSA with the CPAP machine [3].

B. Physiology

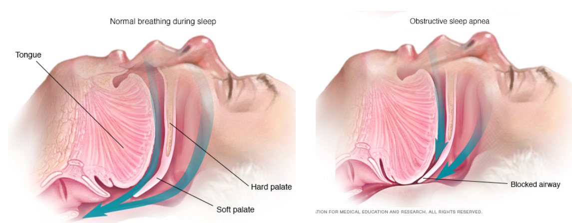

Fig. 1. The Physiology Behind Obstructive Sleep Apnea [4].

OSA occurs when the muscles in the back of the throat relax, causing the soft palate to collapse and block the airway [5], as seen in the above figure. Blocking this region of the throat causes nasal passages to be blocked, preventing the patient from breathing normally.

C. Solution Overview

The goal is to design a small device that tracks breathing patterns in sleep for patients with mild sleep apnea to gauge the quality of their sleep and determine with their doctor their ideal treatment (e.g., if the user would need to continue treatment with the CPAP). This device should alert the user if long stretches of non-breathing occurred, ideally to wake them the choking sensation would. This detection is done primarily by a thermistor, processed by two 555 timers and logic gate to determine whether the alarm needs to be triggered.

While devices already exist to diagnose OSA [1], this device aims to be used nightly, log how many times the user stops breathing, and has an alarm to wake the user before being woken up by the uncomfortable choking and gasping associated with OSA. Future additions to the solution may be added, including a log of breathing patterns and an EEG attachment, but are not included in the current design.

II. Methods

A. Circuit Design

The circuit has several components that work together to provide the signal and analyze it. The first key component is a thermistor circuit which utilizes resistors. One of these changes its resistance in concert with the change in temperature. This creates a sinusoidal signal of voltage which decreases as temperature increases and vice versa. Because this device would be placed underneath the user's nose while they sleep, an exhale (of hot air onto the thermistor) is associated with the temperature increasing and an inhale is associated with temperature decreasing.

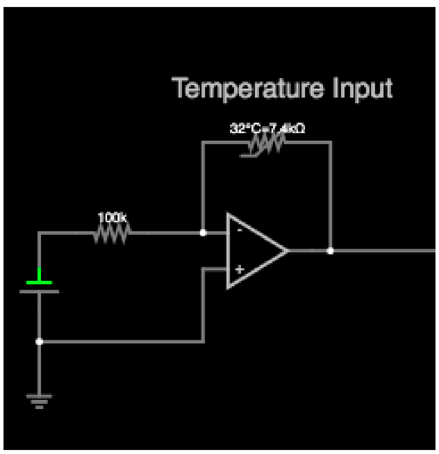

Fig. 2. Thermistor circuit diagram transforms temperature signal into a voltage signal

The next part of the circuit is the hysteretic comparator. This component takes in the sinusoidal output from the thermistor section and compares it to the non-inverting input of the comparator. When choosing the resistors' values, it was dependent on the value wanted at the non-inverting input which is nothing more than a voltage divider. By choosing a value of 0.788V, the midpoint of the expected output voltages from the user, this would allow the comparator to output a square wave of correct spacing of time. This means that when the output signal from the user is above and moving away and back toward 0.788V (i.e. they are finishing their inhale and beginning to exhale) then the output is 0V. This is essentially adding half of breathing in and breathing out to represent the time of a breath in or out. When the output signal is below 0.788V, the signal is at 9V. This creates a square wave signal where the signal is low during the latter half of the inhalation and the first half of the exhalation and high for the other part of the breath. Thus, the square wave signal is dependent on the changes in voltage due to breathing.

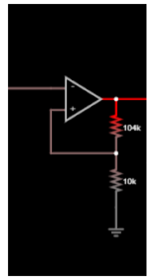

Fig. 3. Hysteretic Comparator

The next key components of the circuit are the In and the Out timers, which are both monostable 555 timers. The signal at the In timer matches the voltage signal that results from the hysteretic comparator. When this signal is low, the person is considered to be breathing in, and while the signal is high the person is considered to be breathing out. The Out signal is the opposite, meaning that when it is low, the person is considered to be breathing out. The two outputs from each timer connect at a NOR gate, determining whether or not the alarm in the next stage of the circuit is triggered. The alarm should trigger when the person is in either phase for over 90 seconds which is considered an abnormal amount of time to not be breathing [6]. The resistances and capacitances here correspond to this time limit. See the equations section for further information on how these values were found.

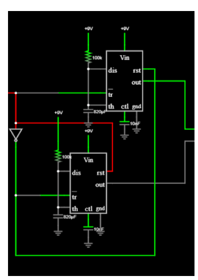

Fig. 4. In and Out Timers. The bottom timer is the Out timer

The final phase is the astable timer which is triggered when the person is on the in phase or on the out phase of their breath for around 90 seconds. The resistors and capacitors here correspond to the frequency of the alarm. This is the output of the circuit. See the equation section for further information.

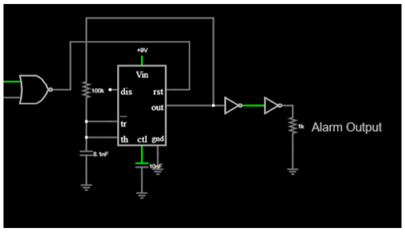

Fig. 5. Astable timer

B. *Equations & Variables:*

For an NTC thermistor, its resistance at a given temperature can be found using the equation

$$R_T = R_o e^{\beta(\frac{1}{T} - \frac{1}{T_o})} \quad (1)$$

In the above equation, R_o and T_o are the thermistor's resistance and standard temperature respectively. At a value of 298.15 K, R_o is 10 kΩ. Using a particular NTC thermistor, the β value is 3977 [7]. Lastly, T is the temperature that the thermistor is experiencing at any given time.

In the first stage of our circuit, the voltage expected to come out of the op amp is as follows

$$V_{out} = \frac{9V \cdot R_T}{100k\Omega} \quad (2)$$

It should be noted that 100kΩ is the value that was chosen for the first resistor. 9V was chosen as the power supply.

The hysteretic comparator obeys the following equation. The hysteretic voltage is set to .788 V which is the middle of the voltage range that we expect. R_3 is considered to be the resistor that is closer to the ground. V_{cc} is set at 9V because that is closed circuit voltage which is the same as the battery voltage.

$$V_{hyst} = \frac{R_3}{R_3 + R_4} Vcc \quad (3)$$

In the In and Out timer stage, the time threshold for when the alarm would sound and the resistance and capacitance values are related by the following equation. The time we chose was 90 seconds.

$$T = 1.1RC \quad (4)$$

In the last stage of the circuit, the pitch of the alarm was set to be 880 Hz. When choosing a resistance value of 100 kΩ, the capacitance value can be found using

$$f_{buzzer} = \frac{1}{1.4 R_b C_b} \quad (5)$$

III. Results

Because the circuit simulation was not working as intended, this results section shows the simulation broken into separate parts, showing the expected signals for each section. It should be noted, with the small time scale and the inability to fully simulate the sinusoid that would be produced from real breathing data, this analysis is done with anticipated results from user data. For normal breathing, an exhaled breath is approximately 33.2 ± 2.3 °C [8] and an inhaled breath is assumed to be approaching room temperature (20 °C), correlating to low voltage outputs at high temperatures and high outputs at low temperatures. This breathing temperature, along with equation 1, is the basis for the minimum and maximum values in figures 6 and 7.

The voltage output from the thermistor component over time of a normal, unobstructed breathing pattern was anticipated to be a sinusoid, illustrated roughly in fig. 6 below where the voltage output is translated from the temperature detected by the thermistor in the device. When the user first breathes in, voltage peaks and the voltage reaches its lowest point when the user breathes out.

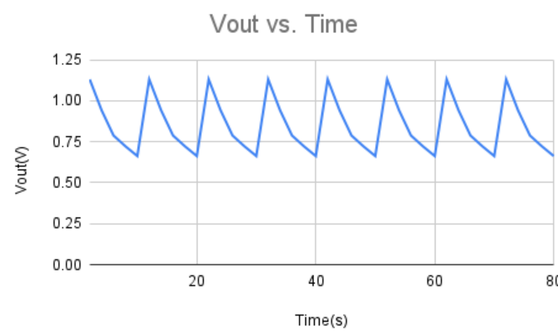

Fig. 6. Expected V_{out} $vs.\ Time$ Plot

Fig. 7 was a representation of an obstruction in the breathing pattern of a user where the flat line represents no change in the voltage output which means there was no change in the temperature. This indicates that the user is not breathing in the flat line of Fig. 7. Whether this is significant or not depends on how long the obstruction occurs. As mentioned above, an obstruction of over 90 seconds would be considered dangerous enough to sound an alarm to alert the user of the obstruction.

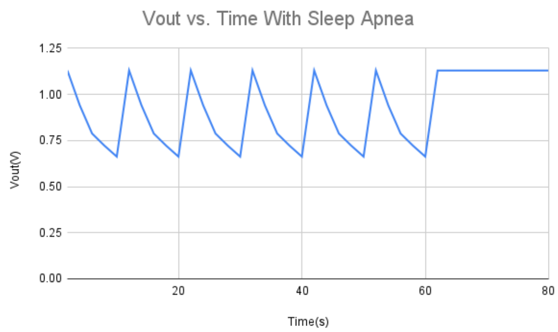

Fig. 7. Expected V_{out} $vs.$ $Time$ Plot for Sleep Apnea Patients

The hysteric comparator was simulated as an AC voltage source with a square wave for a case where the user is breathing normally. In turn, this causes the two 555 timers to alternate high and low, depending on if the comparator output is high or low, which is desired. In turn, this simulates no output from the alarm, which is expected for normal breathing patterns.

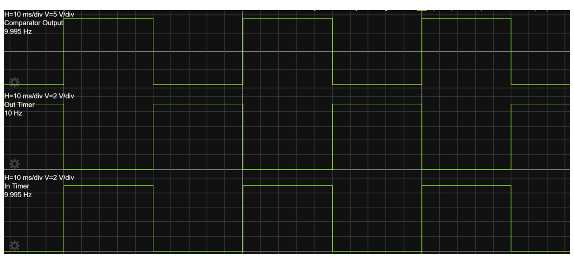

Fig. 8. Expected signal from the hysteretic comparator and in/out timers of the circuit simulation

IV. Discussion

A. Analysis of Results

The most relevant result from the circuit simulation was the component in Fig. 8 above. Although the simulation didn't produce the results we expected, it was edited in such a way that it was able to produce results similar to those we expected.

Because the simulation was not running as intended despite much effort, it was decided upon to break the circuit by stages. The first stage of the circuit had an output with values that were expected as shown in Figure 6. With issues after this stage, it was decided to place a sinusoidal wave into the comparator. After this, a square wave was used to represent the output of the comparator and then sent into the timers to get Figure 8. This figure shows that our In and Out timers work correctly as the signals for these two are mirrors of each other such that when one is high the other is low. This means that if we were to get the other aspects of the circuit to work correctly then our signal analysis would fit correctly with it.

The primary limitation encountered with the circuit simulation was that although the circuit components were valid individually, the simulation didn't produce the voltage output vs. time sinusoidal plot with a square signal we expected. The problem could reside in the temperature fluctuation being manual in the simulator and not accurately representing a natural fluctuation that would be detected by the device.

Another issue faced was finding an appropriate time scale in which to simulate the circuit. The circuit simulation would either be too fast or too slow causing the responses from our timers to range wildly depending on the chosen time step, simulation speed, and current speed. This issue however does not demonstrate a flaw within the circuit's design but rather simply required more calibration from the program that combined with our inability to automate

B. Conclusion

Monitoring sleep apnea is a useful tool for patients and their physicians as it gives the medical professionals a fuller picture of what the patient is going through at night. While current devices like CPAP machines can help a user treat their sleep apnea, it doesn't relay any information of the progression of the affliction.

Through analysis of the breathing waveform when asleep, the proposed device is able to aid in the treatment but also in the recording of the progression of the condition. This was done through the use of thermistors, hysteretic comparators, and timers, in order to analyze the nature of the user's breathing while asleep and determine when dangerously long stretches of apnea occur. Though we were unable to truly simulate our entire design, we were able to see that the relationship between the In timer, Out timer, and comparator signals functions generally as intended showing that our signal processing was generally successful.

One advantage of our design is that we are utilizing a signal dependent on temperature. This signal is likely to have less noise affecting it in comparison to a signal like sound which there are many other sources of while a person is sleeping. On the other hand our signal of temperature is only affected by the inhalation and exhalation of the user and the temperature of the room which is around the same as the inhalation temperature.

There are several extensions of this design that could prove relevant in a medical or bioengineering context. For example, the integration of an EEG would allow for the measurement of brain waves along with the breathing timer. The device could keep a log of both the EEG signals and the breathing as measured by the circuit above. The integration of such a device would allow for a greater understanding of the effect of sleep apnea on brain activity patterns throughout the night which would allow tracking of how the OSA affects the users' sleep quality. This type of understanding of sleep is essential in the increasingly digitized modern world.

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VI. References

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