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Apnoea Monitor Using a Piezo Sensor and Filters

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Abstract: *Sleep apnea syndrome is a prevalent chronic condition that disrupts daily functioning. As its incidence rises, there is a clear demand for an affordable, portable device capable of detecting apneic events and monitoring heart rate. This study followed a three-stage process of design, simulation and implementation, and benchmarked heart-rate data against those obtained from a Ticwatch 3 Pro. Two heart-rate monitor prototypes and one respiration monitor were developed. The first heart-rate prototype employed a low-pass filter with a wrist-mounted piezoelectric sensor. The second prototype used a band-pass filter with the sensor positioned on the tester's back, shared with the respiration monitor. Linear regression and correlation analyses demonstrated that the low-pass filtered wrist-mounted configuration yielded more reliable measurements than the band-pass filtered design.*

Keywords: *Heart Rate Detection, Low-Pass and Band-Pass Filters, Signal Amplification*

1. Introduction

1.1 Overview

Sleep apnea syndrome (SAS) is characterized by cessation of breathing during sleep and is regarded as a common chronic condition (Heatley et al., 2013)[7]. Patients experience repeated upper airway obstruction lasting ten seconds or more (Khattak et al., 2018), manifested by loud snoring, body jerking, or arm fluttering[10]. SAS is associated with sleep fragmentation, daytime somnolence, fatigue, and cardiac irregularities, impairing daily functioning and posing life-threatening risks. Epidemiological evidence links SAS to these health issues (Levy, Ryan, Oldenburg and Parati, 2013)[11]. Treatment options range from lifestyle interventions such as weight loss and exercise for mild cases to surgical procedures for severe apnea (BETTEGA et al., 2000)[4]. Many patients remain unaware of their condition because symptoms occur during sleep.

1.2 Motivation

The disease is then difficult to detect by the patient. Most of the methods used to diagnose SAS are traditional, such as EEG, EMG, EEG and airflow velocity during sleep (Jacob et al., 1995)[8]. Testing by traditional methods requires a lot of medical resources. And it is inappropriate for a common chronic condition like apnea syndrome. Because 936 million (30-69 years old) adults worldwide suffer from mild to severe obstructive sleep apnea syndrome, and 425 million adults suffer from moderate to severe obstructive sleep apnea syndrome (Benjafield et al., 2019)[3]. For the average person they need an inexpensive and portable device to assist them in determining if they have SAS.

1.3 Objectives

The goal of this project is to develop an apnea monitor with advantages in price and portability. Many smart wearable devices have emerged in recent years. These devices include bracelets, watches, belts, armbands, etc.. But most of them can only detect heart rate, blood oxygen and noise. There are almost no devices for detecting apnea syndrome. This project will develop a low-cost and portable monitor for SAS. The function of this monitor is not only to support apnea monitoring but also to support heart rate monitoring.

2. Literature Review

The objective of this project is to develop a breathing and heart rate monitor of reasonable price and

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portability. This chapter will study the methods used by predecessors in this field. A scheme that meets the project's needs is obtained by analyzing the advantages and disadvantages of the methods proposed by predecessors.

2.1 Breath signal acquisition

Respiratory airflow generates characteristic sounds as air rubs against the nasal or oral walls. Exploiting this, (Alqassim et al., 2012) proposed real-time sampling of breathing sounds via a smartphone microphone[1]. Their algorithm estimates breath frequency and amplitude, and flags apnea when no signal is detected for ten seconds or more. This approach is low-cost and portable—requiring only a ubiquitous smartphone—but its accuracy deteriorates in noisy environments, as ambient sounds can obscure the respiratory signal.

Exhaled gas temperature differs from ambient air; consequently, (Fei, Pavlidis and Murthy, 2009) used infrared imaging to capture facial thermal signals and detect apnea[5]. However, this method requires costly equipment (infrared cameras cost approximately £50), depends on unobstructed facial exposure to maintain signal fidelity, and becomes unreliable when exhaled air temperature is close to ambient levels, since breath temperature is about 34 degrees Celsius (Anghel and Iacobescu, 2013)[2].

2.2 Heartbeat signal acquisition

The acquisition method described above explores how to obtain the respiratory signal of a person during sleep. However, this project also requires the measurement of a person's respiratory signal. The heartbeat signal is usually represented in the form of vibrations on the human epidermis. (Fujita et al., 2012) proposed the use of a PVDF piezoelectric sensor to capture the heartbeat signal[6].The sensor is in the form of a thin film that fits perfectly to the human skin. The PVDF sensor acquires the signal in a similar way to a fibre optic sensor. They both need to be fixed to the person's chest. This type of measurement fixed to the chest is uncomfortable.

For this project, a ceramic piezoelectric sensor will be used. This sensor costs less than £1 and meets the low-cost requirement. It is ideal for this project as it can capture not only the heartbeat but also the breathing signal.

2.3 Microprocessors

The microprocessor is the heart of the whole project. It is responsible for processing the signals sampled by the sensors, calculating the heart rate and alerting to pauses in breathing. (Rotariu et al., 2016) used the Arduino Leonardo in his study and MCU he chose was the ATmega32u4[14]. (Rasheed et al., 2019) used the msp430f149 microcontroller from Texas Instruments in his study[12]. The performance of the Msp430f149 is higher than that of the ATmega32u4. However, the ATmega32u4 is easier to develop based on the Arduino platform because Arduino provides IDE and a large number of libraries for developers. The Arduino-based ATmega328P microprocessor was chosen for this project because the project is in the research phase, and there is no rush to mass production.

2.4 Filters

The project's filter architecture mitigates both high- and low-frequency interference. (Ren et al., 2015)employed a conventional analogue low-pass filter[13]. (Rotariu et al., 2016) proposed a digital band-pass filter for respiratory signals[14]. Analogue filters were selected for their ease of implementation and independence from microprocessor resources. Band-pass filters, which attenuate both low- and high-frequency components, are inherently more complex than low-pass filters, which remove only high-frequency noise. Accordingly, three prototypes were developed: a wrist-mounted heart-rate monitor using a low-pass filter; a back-mounted heart-rate monitor using a band-pass filter to suppress both respiratory (low-frequency) and other noise; and a respiration monitor employing a low-pass filter to reject frequencies above the respiratory band.

3. Methodology

The primary objective of this project is to detect sleep apnea episodes, trigger an alert, and record and display the patient's heart rate. Based on prior studies, heartbeat signals are first acquired from the wrist

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and respiratory signals from the back, each processed with a low-pass filter to remove high-frequency noise. Subsequently, a band-pass filter is evaluated for back-mounted heartbeat detection to compare signal quality between wrist and back placement and between low-pass and band-pass filtering approaches.

Beyond filter development, two additional objectives are addressed. First, the sensor's weak analogue output is amplified to ensure accurate microprocessor sampling. Second, a microprocessor system is implemented to issue apnea alerts and display heart-rate data in real time.

3.1 Research design

The research employs an experimental methodology, encompassing circuit simulation and physical implementation. Simulation is pivotal, as it accelerates the development cycle via rapid design verification, reduces implementation risk through early fault detection, and lowers costs by using virtual components.

3.1.1 Amplifier

The amplifier is implemented in two steps. The first step is to design a Multisim circuit based on the desired gain multiplier. The amplifier is then simulated on Multisim. Finally, the results are analysed. After a successful simulation, the actual circuit is implemented on a breadboard.

3.1.2 Filters

There are two types of filters to be implemented in this project, the former is a low-pass filter, and the latter is a bandpass filter. The design of the filters is more complex than that of the amplifiers, so the filter design tools provided by Texas Instruments and analogue devices were used to design the filters required for this project. Firstly, the user has to set the parameters of the filter, such as the passband frequency and the stopband frequency of the low-pass filter. The design tool then generates a filter that matches the parameters. Next, the filter is simulated on Multisim. A bode plotter and an oscilloscope are used in Multisim to check that the filter meets the design requirements. Finally, the circuit that meets the design requirements is implemented on a breadboard.

3.1.3 Microprocessor

The microprocessor needs to determine if the tester is experiencing apnoea and display the heart rate on the screen. Arduino programming is divided into the following steps. Firstly, each step of the program needs to be represented as a flowchart. The flowchart is an important step for microcontroller programming. It is the first draft of the programming. The flowchart makes the programmer's programming clear. It increases the speed at which the programmer can program. The program is then written for the Arduino according to the flowchart. Then use Arduino IDE to compile the code of MCU. The next step is to upload the compiled binary file to the Arduino MCU in Proteus and then simulate the program. The last step is to burn the program verified by the simulation to Arduino UNO.

3.2 Data sampling

The project dataset comprises respiratory and cardiac waveforms as well as quantitative heartbeat counts. Waveforms were acquired using an oscilloscope (which can export data as images or CSV tables) and via the Arduino IDE's serial plotter, which renders real-time values from 0 to 1023 corresponding to a 0-5 V analogue input. Heartbeat counts are obtained by timing ten successive cardiac cycles, and beats per minute are computed from the elapsed interval.

4. Result

In this chapter, the results of the development of amplifiers, filters and microprocessors are first presented. The simulation section includes the circuit diagrams they have simulated and the results of the simulation. The real part is the circuit diagram implemented on the breadboard after the simulation. After the implementation of the first version of the heart rate monitor, an apnea monitor and a second version of the heart rate monitor was developed for this project. Then the heart rate results of the heart rate monitor developed in this project will be compared to those of the ticwatch3 as a way to verify the accuracy of the heart rate monitoring in this project. Finally, this project compares the difference between heart rate monitoring on the wrist and heart rate monitoring on the back and the difference between a low pass filter and a bandpass filter. The first version is shown in Figure 1.

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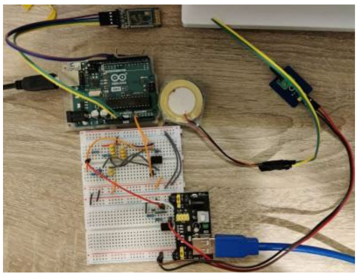

Figure 1: The first edition of heartbeat monitor

4.1 First version of the heart rate monitor

In this section, an amplifier is first developed, followed by a filter and finally the development of the Arduino.

4.1.1 Amplifiers

The first step is to design a ten times gain co directional amplifier on Multisim. Figure 2 shows the simulation circuit diagram from Multisim.

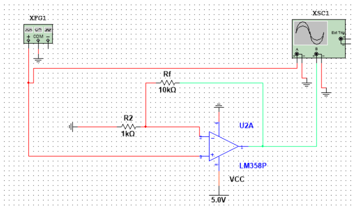

Figure 2: Amplifier circuit

The second step is to simulate the op-amp. The original waveform in Figure 3 is 2Hz, 400mVp. The original waveform passes through the amplifier and becomes 3.5V. The reason there is no reverse voltage in the figure is that the negative side of the amplifier is connected to a 0v ground. The reason for this is that Arduino cannot provide reverse voltage. The experiment found that the heartbeat waveform determines the human heartbeat, and the voltage of this waveform is positive.

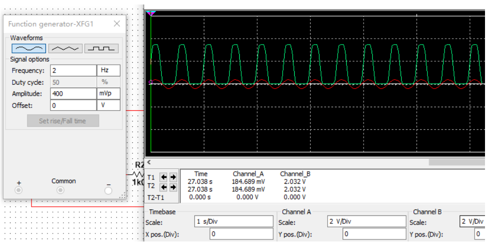

Figure 3: Simulation of the amplifier

The third step is to implement the real circuit on a breadboard. Figure 4 below shows a real circuit implemented on a breadboard. Figure 5 below is a screenshot of the oscilloscope output, which shows the amplification of the real circuit. Due to the influence of environment, amplifier performance, and component accuracy, the magnification in the actual circuit is 5.1 times.

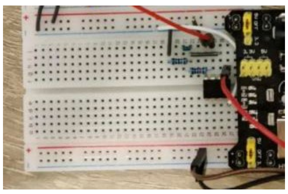

Figure 4: Amplifier real circuit

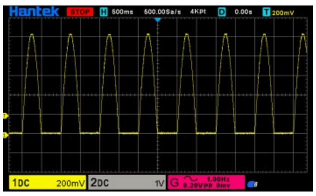

Figure 5: Oscilloscope output of amplifier

4.1.2 Low-pass filter

Figure 6 shows the circuit diagram for the low-pass filter design in Multisim. Figure 7 shows the low-pass filter bode plot. It can be seen that the filter has a cut-off frequency of 2.7 Hz and that the signal is

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attenuated by -40db (100 times) when the frequency reaches 6.3 Hz. The simulation in Figure 8 shows that significant attenuation occurs when the signal reaches 5hz. Figure 9 shows the actual circuit diagram of this low pass filter. Figure 10 shows that there is a significant attenuation in the actual circuit when the signal reaches 4Hz. The actual circuit is in general agreement with the results of the simulation. This shows that the filter design is feasible.

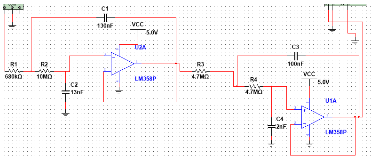

Figure 6: Low-pass filter circuit in Multisim

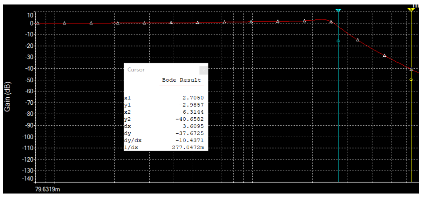

Figure 7: Low-pass filter bode plot

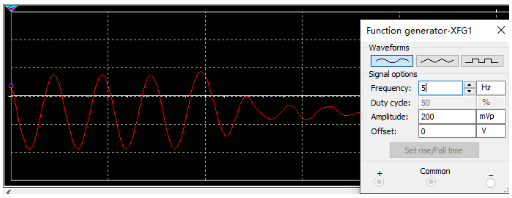

Figure 8: Simulation of the low-pass filter

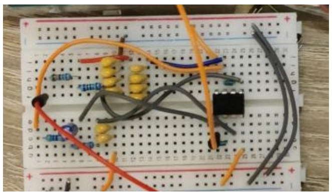

Figure 9: Low-pass filter real circuit

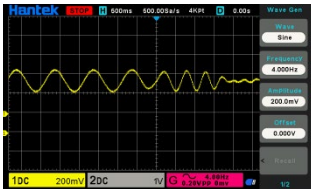

Figure 10: Oscilloscope output of low-pass filter

4.1.3 Arduino

The Arduino development comprised three key stages. First, the system measured ten consecutive heartbeats to calculate heart rate. As shown in Figure 11, a switch simulates detection of the rising and falling phases of the heartbeat waveform; each detected cycle increments the timer until ten beats are recorded, from which heart rate is derived. Second, an Android application was implemented using MIT App Inventor to control monitoring and receive data. Figure 12 depicts the Arduino UNO with HC-05 Bluetooth module, and Figure 13 shows the app interface. The user pairs with HC-05, clicks on the Start button, and the app issues a Bluetooth command to begin measurement. Finally, the Arduino computes the heart rate and transmits the result back to the app for display.

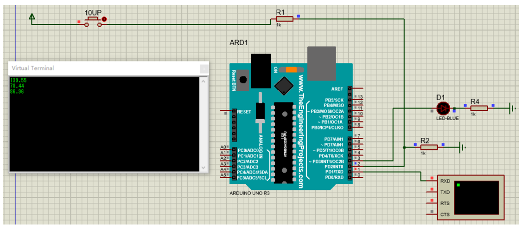

Figure 11: simulation of Arduino in Proteus

Figure 12: UNO with the Bluetooth module

Figure 13: Application interface

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4.2 Breath monitoring

Like heartbeat monitoring, this part also develops amplifiers, low-pass filters and Arduino. The amplifier is the same as the previous heartbeat monitoring amplifier, which is a ten times gain in the same direction.

4.2.1 Low-pass filter

Figure 14 shows the low-pass filter circuit in Multisim, and Figure 15 its gain plot, which has a 6 dB passband gain and a 0.5 Hz cutoff. At 1 Hz the signal is attenuated by 40 dB (Figure 16), effectively removing heartbeat interference from the respiratory signal. Since human respiration is 0.2–0.33 Hz and heartbeats are 1–1.66 Hz (John Camm and Fei, 1996)[9], this filter design isolates breathing. Figure 17 is the real-world circuit, and Figure 18 confirms that its performance matches the simulation, attenuating frequencies above 1 Hz.

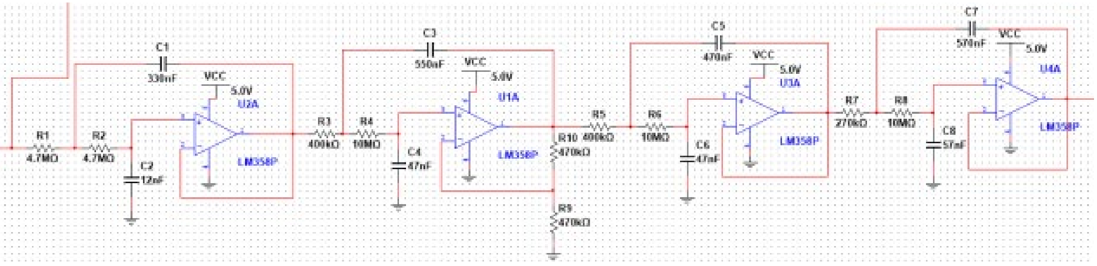


Figure 14: Low-pass filter circuit in Multisim

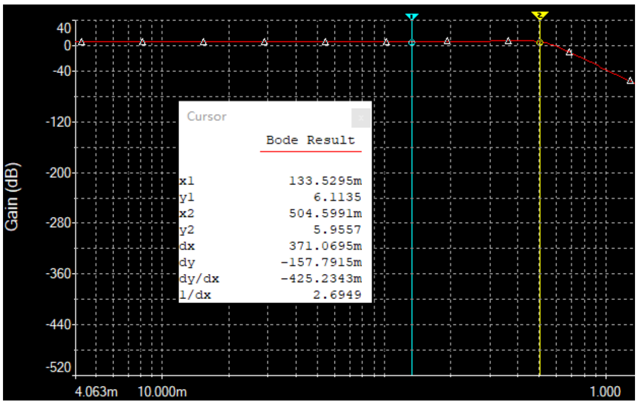
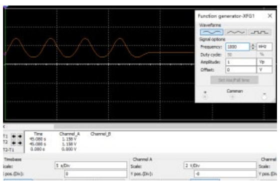


Figure 15: Low-pass filter bode plot



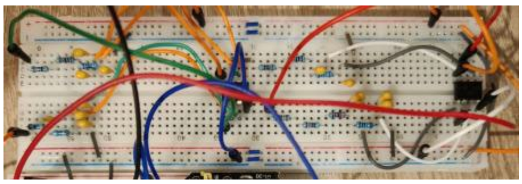


Figure 16: Simulation of the low-pass filter *Figure 17: Low-pass filter real circuit*

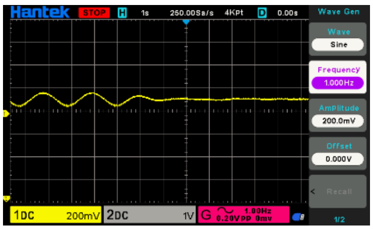


Figure 18: Oscilloscope output of low-pass filter

4.2.2 Arduino

The Arduino nano is used as the MCU for this part of the respiratory monitoring. The performance of the nano and uno is identical. The difference between them is the size of the board and the number of interfaces. The nano is used in the respiratory monitoring section to make experimentation easier. As we

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are currently in the experimental phase, respiratory monitoring and heartbeat detection are implemented with uno and nano respectively.

Figure 19 shows the nano, the amplifier, and the alarm led light. The Arduino sketch continuously reads the sensor voltage on the analog input (0 - 5 V), which the ADC converts into a 0 - 1023 integer. Each reading is offset by a constant CHECK and stored in d. When d ≥ 40, the code recognizes the rising phase of the breath waveform; when d ≤ 10, it recognizes the falling phase. Detecting a rising event followed by a falling event signals one complete breath, and the timestamp is saved in times1. In each loop iteration, the current time (times2) minus times1 is computed as t. If t exceeds 10 seconds, a respiratory pause is detected and the alarm LED is turned on. As soon as the next breath is detected, the LED is switched off.

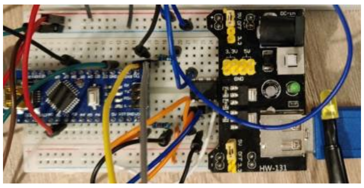

Figure 19: The real circuit of the respiratory monitor

4.3 Second edition heart rate monitor

In this section, the use of bandpass filters to obtain a heartbeat signal from a person's back is investigated. The advantage of using this method is that only one sensor is needed to get the respiratory signal and the heartbeat signal. Only a low-pass filter is needed to remove more than 1hz of the signal for the respiratory signal. For the heartbeat signal, however, not only the respiratory signal below 0.5 Hz has to be removed but also the high-frequency signal above 10 Hz. The disadvantage of measuring the heartbeat from the back is that both high and low frequency signals have to be removed. Also, the signal strength is lower on the back compared to the wrist. The second version of the heart rate monitor uses the same piezoelectric sensor as the respiratory monitor. For this project this sensor is fixed to a board. The measurement requires the tester to lie on the plate. A picture of it is shown in Figure 20.

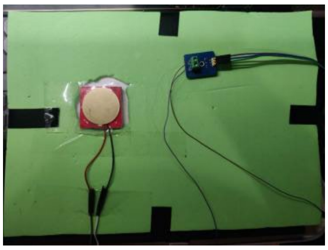

Figure 20: Piezoelectric sensor of the second edition of heartbeat and respiration monitor

4.3.1 Amplifiers

The amplifiers used in this section are the same as those designed earlier.

4.3.2 Filters

Figure 21 shows a circuit diagram of a bandpass filter in Multisim. Figure 22 shows the gain plot for this filter. The gain plot shows that the filter can attenuate up to -20db at 0.4Hz and its passband range of 1Hz-2.2Hz. Figure 23 and 24 show the simulation of this filter. Figure 23 shows the signal changing from 1Hz to 0.4Hz, and Figure 24 shows the signal changing from 2.2Hz to 6Hz. After their change in frequency, a significant attenuation of the signal can be seen. Figure 25 shows a real circuit of this filter. Figure 26 and Figure 27 show the test results of the actual circuit. Their results match the Multisim simulation. The filter in the actual circuit removes signals with frequencies above 6hz and below 0.5Hz.

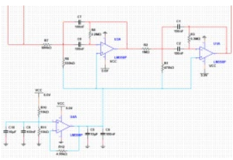

Figure 21: bandpass filter circuit in Multisim

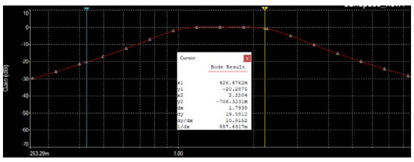

Figure 22: bandpass filter bode plot

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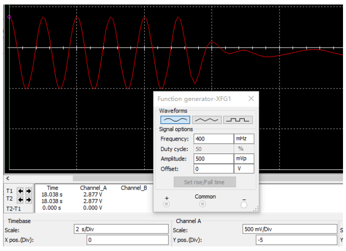

Figure 23: Bandpass filter (1Hz to 0.4Hz)

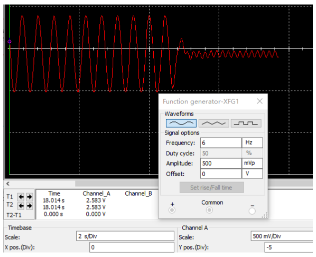

Figure 24: Bandpass filter (2.2Hz to 6Hz)

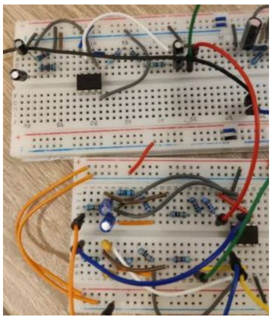

Figure 25: bandpass filter real circuit

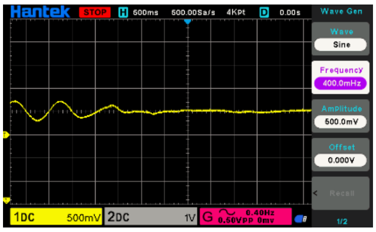

Figure 26: Oscilloscope output (1Hz to 0.5Hz)

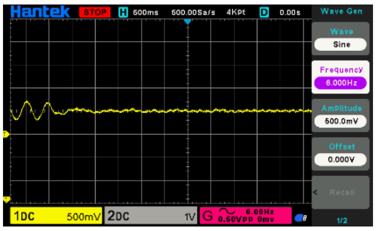

Figure 27: Oscilloscope output (2.2Hz to 6Hz)

4.3.3 Arduino

The code for this section is essentially the same as that used for the wrist measurement. The changing part is in the judgment of the heartbeat waveform, and its judgment logic is the same as that of the respiratory signal.

4.4 Data and results

Table 1 shows two sets of data for heart rate measurements. The first set of data is from the Ticwatch. The second set of data is from the first version of the heart rate monitor. These two sets of data were sampled simultaneously and with the same environmental factors.

Table 1: Data for heart rate (first version)

Type	1	2	3	4	5	6	7	8	9	10
Ref(ticwatch)	86	85	81	77	80	78	76	76	75	81
Experiment	89	88	93	84	80	80	80	78	78	86

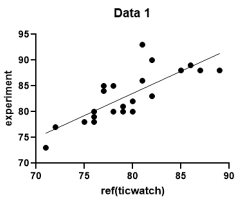

Figure 28: Linear regression

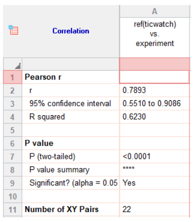

Figure 29: Correlation

Next, this next project performed a linear regression analysis and a Pearson correlation analysis on the two sets of data. The results of their analysis are shown in Figure 28 and Figure 29. It can be concluded

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that the version one measurements are positively correlated with the Ticwatch measurements. Moreover, the Pearson correlation r was 0.7893. This value is very close to 1, indicating that the results of version one are strongly correlated with the results of the Ticwatch. This then further suggests that the heart rate measurements of version one are reliable.

The first set of data in Table 2 is from Ticwatch and the second set of data is the heart rate data measured by the second version of the heartbeat monitor.

Table 2: Data for heart rate (second version)

Type	1	2	3	4	5	6	7	8	9	10
Ref(tiwatch)	86	85	83	85	83	83	80	84	79	75
Version2	89	81	82	93	90	86	76	72	72	73

As with the first version, this project performed a linear regression analysis and correlation between the two new sets of data. The results of their analysis are shown in Figure 30 and Figure 31. It was concluded that the second version of the heartbeat monitor had an r of only 0.6454. It did not have as high a correlation as version 1. This indicates that version 2 is not as reliable as version 1. The reason for this result is that the piezoelectric sensor of version 2 is placed on the back of the tester. The movement of the tester's body can affect the results of the measurement.

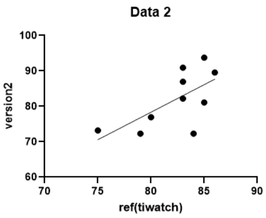
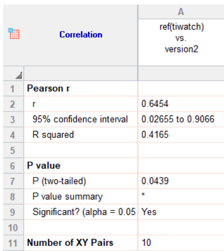


Figure 30: Linear regression



	Correlation	A ref(tiwatch) vs. version2
1	**Pearson r**	
2	r	0.6454
3	95% confidence interval	0.02655 to 0.9066
4	R squared	0.4165
5		
6	**P value**	
7	P (two-tailed)	0.0439
8	P value summary	*
9	Significant? (alpha = 0.05	Yes
10		
11	**Number of XY Pairs**	10

Figure 31: Correlation

The cost of this project is lower than that of (Alqassim et al., 2012) and (Fujita et al., 2012) [1][6] because it employs ceramic piezoelectric sensors priced at £1 each, in contrast to the more expensive infrared cameras, Kinect v2 devices, PVDF sensors and UWB radars. Sensor and microprocessor dimensions meet portability requirements. Unlike (Alqassim et al., 2012), this system provides both respiratory and heart-rate monitoring, and it is more compact than the solution of (Fei, Pavlidis and Murthy, 2009), since the ceramic sensor weighs under 10 grams while infrared cameras, Kinect v2 units and UWB radars exceed that weight[1][5]. Moreover, image- or reflection-based approaches necessitate high-performance computers, whereas the present design runs on a single Arduino.

5. Conclusion and Limitatons

5.1 Conclusion

The project begins with a background review and identifies user requirements for an affordable, portable breathing and heart-rate monitor. By comparing prior work, suitable sensors, filters, and microprocessors were selected. The development followed a sequence of design, simulation, and physical prototyping, during which breathing and heartbeat waveforms were recorded via oscilloscope or the Arduino IDE. Finally, both the simulated and physical circuit diagrams are presented, and the heart-rate measurements from two prototype versions are compared against those of a Ticwatch.

5.2 Achievements and limitations

The first prototype placed the sensor on the tester’ s wrist and used only a low-pass filter, yielding highly reliable heart-rate readings comparable to a Ticwatch. The second prototype, mounted on the tester’ s back, required a band-pass filter to isolate the heartbeat from respiratory and high-frequency noise, resulting in less stable measurements. A respiration monitor using the same piezoelectric sensor was also developed, with separate low- and band-pass filters for breathing and heartbeat signals. Currently, heart-rate and respiration monitoring run on separate microprocessors, increasing cost, and rely on analogue filters whose component tolerances can introduce measurement errors.

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5.3 Further research

Future work may address two design limitations. First, integrating respiration and heart-rate monitoring on a single microprocessor would require using its interrupt functionality to switch between tasks rapidly, which demands a solid understanding of interrupt handling. Second, replacing analogue filters with digital filters would eliminate component-tolerance errors but requires expertise in digital filter design.

References

[1] Alqassim, S., Ganesh, M., Khoja, S., Zaidi, M., Aloul, F. and Sagahyroon, A. (2012). Sleep Apnea Monitoring using mobile phones. 2012 IEEE 14th International Conference on e-Health Networking, Applications and Services (Healthcom).

[2] Anghel, M.-A. and Iacobescu, F. (2013). The Influence of Temperature and CO2 in Exhaled Breath. 16th International Congress of Metrology. [online] Available at: https://cfmetrologie.edpsciences.org/articles/metrology/pdf/2013/01/metrology_metr2013_10012.pdf [Accessed 14 Feb. 2021].

[3] Benjafield, A.V., Ayas, N.T., Eastwood, P.R., Heinzer, R., Ip, M.S.M., Morrell, M.J., Nunez, C.M., Patel, S.R., Penzel, T., Pépin, J.-L., Peppard, P.E., Sinha, S., Tufik, S., Valentine, K. and Malhotra, A. (2019). Estimation of the global prevalence and burden of obstructive sleep apnoea: a literature-based analysis. The Lancet Respiratory Medicine, [online] 7(8), pp.687–698. Available at: https://www.thelancet.com/journals/lanres/article/PIIS2213-2600(19)30198-5/fulltext.

[4] BETTEGA, G., PÉPIN, J.-L., VEALE, D., DESCHAUX, C., RAPHAËL, B. and LÉVY, P. (2000). Obstructive Sleep Apnea Syndrome. American Journal of Respiratory and Critical Care Medicine, 162(2), pp.641–649.

[5] Fei, J., Pavlidis, I. and Murthy, J. (2009). Thermal Vision for Sleep Apnea Monitoring. Medical Image Computing and Computer-Assisted Intervention – MICCAI 2009, pp.1084–1091.

[6] Fujita, T., Shiono, S., Kanda, K., Maenaka, K., Hamada, H. and Higuchi, K. (2012). Flexible Sensor for Human Monitoring System by Using P(VDF/TrFE) Thin Film. 2012 Fifth International Conference on Emerging Trends in Engineering and Technology.

[7] Heatley, E.M., Harris, M., Battersby, M., McEvoy, R.D., Chai-Coetzer, C.L. and Antic, N.A. (2013). Obstructive sleep apnoea in adults: A common chronic condition in need of a comprehensive chronic condition management approach. Sleep Medicine Reviews, 17(5), pp.349–355.

[8] Jacob, S.V., Morielli, A., Mograss, M.A., Ducharme, F.M., Schloss, M.D. and Brouillette, R.T. (1995). Home testing for pediatric obstructive sleep apnea syndrome secondary to adenotonsillar hypertrophy. Pediatric Pulmonology, 20(4), pp.241–252.

[9] John Camm, A. and Fei, L. (1996). Chronotropic incompetence-part i: Normal regulation of the heart rate. Clinical Cardiology, 19(5), pp.424–428.

[10] Khattak, H.K., Hayat, F., Pamboukian, S.V., Hahn, H.S., Schwartz, B.P. and Stein, P.K. (2018). Obstructive Sleep Apnea in Heart Failure: Review of Prevalence, Treatment with Continuous Positive Airway Pressure, and Prognosis. Texas Heart Institute Journal, [online] 45(3), pp.151–161. Available at: https://www.ncbi.nlm.nih.gov/pmc/articles/PMC6059510/pdf/i1526-6702-45-3-151.pdf.

[11] Levy, P., Ryan, S., Oldenburg, O. and Parati, G. (2013). Sleep apnoea and the heart. European Respiratory Review, 22(129), pp.333–352.

[12] Rasheed, A., Iranmanesh, E., Li, W., Xu, Y., Zhou, Q., Ou, H. and Wang, K. (2019). An Active Self-Driven Piezoelectric Sensor Enabling Real-Time Respiration Monitoring. Sensors, 19(14), p.3241.

[13] Ren, L., Hasan, M.S., Islam, S.K., Fathy, A.E. and Fathy, R.A. (2015). UWB baby and sleep apnea monitor. 2015 IEEE International Symposium on Antennas and Propagation & USNC/URSI National Radio Science Meeting.

[14] Rotariu, C., Cristea, C., Arotaritei, D., Bozomitu, R.G. and Pasarica, A. (2016). Continuous respiratory monitoring device for detection of sleep apnea episodes. 2016 IEEE 22nd International Symposium for Design and Technology in Electronic Packaging (SIITME).