# GM2 Interim Report

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***Abstract*—This report details the initial research and ideas of Team 2 for the creation of a wearable health monitor. This process involved the development of a temperature sensor, heart rate sensor, blood oxygen sensor and blood pressure monitor, with a view to combine the three devices in a wearable 'wrist-watch' design in future work.**

## I. Introduction

Providing adequate health care to communities in remote locations has always been an on-going struggle. By producing a wearable device that can transmit information during a health care emergency, we aim to combat this problem. By measuring human vitals such as heart rate, temperature, blood oxygen and blood pressure, the device will have a clear overview of an individuals health. Combining this data, the device will be able to determine whether the individual is suffering from a medical emergency.

Once an emergency is detected, the device will initially attempt to warn the user by emitting a signal; this will be done using a buzzer or an LED. In addition to the physical signals, the device will incorporate a radio transmitter to inform health care operatives in larger cities. These operatives will then be able to assist as necessary. Furthermore, each device will have a unique serial code related to its region, and there is therefore the potential to spot epidemics before the outbreak becomes too severe.

Due to the time constraints of the project, our aim is to lay the initial foundation. The device has the potential to 'shorten the distance' between remote communities and the relevant care, and we hope that the idea can be taken further through additional research.

## II. The Clever Idea

In order to monitor the vitals of a user in a wearable and transportable format, our product will consist of a temperature sensor, a heart rate and blood oxygen sensor as well as a device to determine blood pressure. The sensors will be connected to a microcontroller - in this case the Arduino Uno - for ease of data collection and processing to determine when the user is unwell. The sensors will be attached to a Velcro strap which will be worn on the lower forearm or wrist region, and the batteries and microcontroller will, for the initial prototype, be placed in a bum bag.

### *A. Temperature Sensor*

Temperature is regarded as the first vital sign as it is very useful in indicating the physiological state of a person; a higher than normal core temperature (greater than 37.5°C) is indicative of a temperature, whilst a lower than normal temperature (less than 35°C) is indicative of circulatory shock, which also has the symptoms of low blood pressure and a weak heart rate [1].

*1) Theory:* It was decided to use a Negative Temperature Coefficient (NTC) thermistor in this project, due to the high accuracy and resolution compared to similarly-priced semiconductor devices. These devices also offer excellent immunity to noise and lead resistance, which is important in a device with multiple sensors and wires [2].

As the temperature increases, the resistance of an NTC thermistor decreases, meaning the temperature can be found by measuring the resistance of the thermistor. However, for an NTC the relationship is highly non-linear, and as such the resistance-temperature characteristic and hence beta parameter of each thermistor was determined experimentally over the expected range of temperatures. The following approximate relationship (the Beta parameter equation) was employed to linearise the thermistor output and translate this into a skin temperature reading:

$$\frac{1}{T} = \frac{1}{T_0} + \frac{1}{\beta} ln\left(\frac{R}{R_0}\right) \tag{1}$$

The variable $T$ is ambient temperature in Kelvin, $T_0$ is room temperature, $\beta$ is the beta constant, $R$ is the thermistor resistance and ambient temperature and $R_0$ is the thermistor resistance at $T_0$.

Since the thermistor sensor was connected to the Arduino microcontroller, the output had to be digitised by the Arduino's analogue-to-digital converter (ADC). As the ADC cannot measure resistance directly, a simple voltage divider circuit (see Appendix A, Figure 4) was used to obtain an output voltage, which was then digitised by the ADC, and could be directly related to the resistance of the thermistor by a simple potential divider equation.

*2) Design:* It was decided that the sensor be placed in contact with the skin on the wrist or lower forearm, as this is best integrated with the rest of the proposed design. It is well understood that the skin temperature of the forearm ($T_{fs}$) tends to be a few degrees lower than that of the core body temperature ($T_c$), and that $T_{fs}$ fluctuates far more with environmental temperature ($T_e$) than $T_c$. [1]

In order to account for these two effects, a multiple linear regression analysis was performed by the team on data (summarised in the Appendix A, Table II) obtained by *Webb, 1992* in his experiment that involved measuring the temperatures of different parts of the body of volunteers in extreme conditions.

A relationship was found relating the auditory canal temperature, used as an approximate core body temperature ($T_c$), the ambient environment temperature ($T_e$) and the forearm skin temperature ($T_{fs}$):

$$T_c = 0.022453T_e + 0.11698T_{fs} + 32.416 \tag{2}$$

This equation was implemented in the code to convert the temperature determined by the NTC thermistor circuit to an approximate core body temperature, with the intention of using this result in order to determine whether or not the temperature of the user is abnormal, and as such whether or not an alarm should be raised. It required the use of two temperature sensors, one to be placed on the outside of the device, to measure $T_e$ and one in contact with the forearm to measure $T_{fs}$

### B. Heart Rate and Blood Oxygen Sensor

There are currently two main methods of measuring heart rate; electrocardiography and photoplethysmography. With the device located on the wrist, the aim is to use both the radial artery and ulnar artery to detect pulse related information.

*1) Theory:* Photoplethysmography (PPG) is a non-invasive pulse measuring technique that is used in a large majority of current wearable fitness trackers. The technology relies on the result that arterial and venous blood interacts with the cardiac, respiratory, and autonomic system, significantly altering the ability of the local tissue to absorb light. It is generally agreed that the cardiac component of the waveform comes from the site of maximum pulsation within the arteriolar vessels, where pulsatile energy is converted to smooth flow just before the level of the capillaries [3]. By measuring the light absorption of local tissue using an LED light source and a light detector, it is possible to determine the blood oxygen content and the heart rate.

Current PPG sensors use two different frequencies of light, generally red ($\approx$ 660nm) and infrared ($\approx$ 940nm). Due to the fact that the infrared light is less influenced by the blood oxygen saturation, it is this frequency that is used to produce a cardiac cycle graph. It is also used as a baseline to calibrate the blood oxygen calculations. Initially, the ratio of red to infrared signals is calculated as shown below in equation 3:

$$Z = \frac{(AC_r/DC_r)}{(AC_{\text{ir}}/DC_{\text{ir}})} \tag{3}$$

Due to the nonlinear nature of the PPG sensors, they must then be calibrated with real blood oxygen values (Sp02). The calibration equation for the MAX30100 sensors is shown below in equation 4:

$$Sp02 = (-45.06 * Z + 30.354) * Z + 94.845 \tag{4}$$

Displayed PPG signals are the inverted infrared waveforms, as shown in Figure 1. It is inverted so that the waveform represents the arterial pressure.

This signal, however, is the result of a series of amplifiers and filters that simplify the complex unfiltered waveform. First, a band pass filter is used to eliminate slow gradual changes, as well as sharp peaks [4]. This acts to smooth the waveform and ensures that it is "auto-centered". Next, amplifiers are used to emphasise the pulsatile component of the signal. This is done in two ways; the light intensity of the LED is continuously adjusted, and so too is the amplification provided by the electronics.

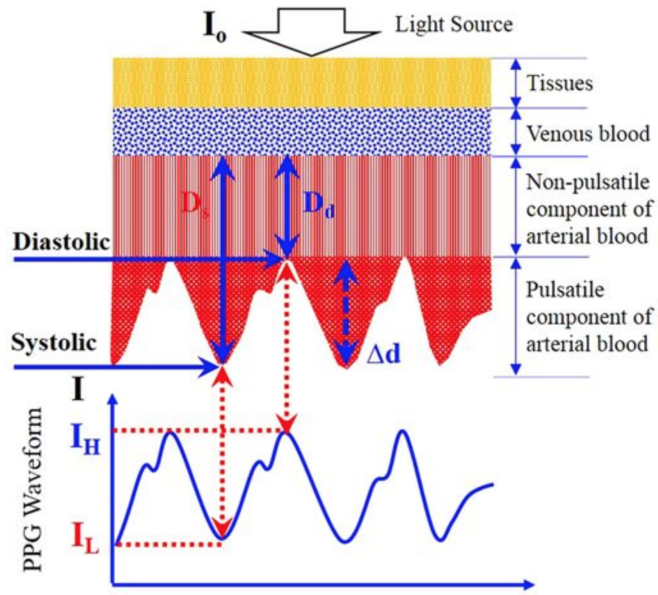



Fig. 1. PPG output plot [5]

A useful feature of PPG that has the potential to be implemented on our device is its sensitivity to heart irregularity, allowing for the measurement of heart rate variability (HRV) [6]. A large HRV is a sign of an underlying cardiac health problem. However, it is worth noting that this technique is still in its early phases, and will potentially be something to consider in future iterations of the device.

Furthermore, PPG technology can be used in conjunction with ECG sensors to calculate the pulse transit time (PTT). This is a measurement of how long it takes a pulse to reach a distant part of the body, and is an indicator of blood pressure. This is explained in full in the Blood Pressure segment of this report.

An indicator of underlying heart rate disease can be linked to the resting heart rate value and the maximum achievable heart rate within a specified time frame. A normal heart rate for adults is expected to be between 60 and 100 beats per minute. The resting heart rate can be estimated by averaging the values recorded over 24h. The maximum achievable heart rate can be estimated with the rule of thumb $85\% * (220 - \text{person age})$.

*2) Design:*

*Arduino setup:* To take heart rate and oxygen level measurements a MAX30100 sensor was connected to an Arduino board. The library Arduino-MAX30100-master provided the required functionality for blood oxygen calculation and heart rate estimation. Every heart beat is detected within the software and a heart rate value in beats per minute is the output.

*MAX30100:* The sensor chosen is an integrated pulse oximetry and heart rate monitor sensor. It combines two LEDs, a photodetector and low-noise analog signal processing to detect pulse oximetry and heart-rate signals. The output of the device is shown in Appendix B in figure 8. The MAX30100 operates from 1.8V and 3.3V power supplies which are provided by the Arduino board.

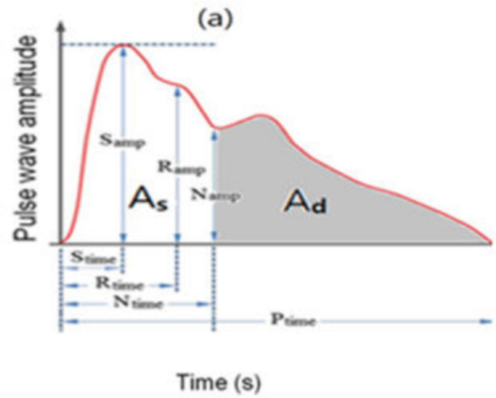
*Attachment to the body:* To achieve maximum accuracy within a wearable healthcare device the positioning of the heart rate sensor is essential. Firstly, a constant pressure needs to be maintained between the device and the patient body. In the prototyping stage the measurements are taken by manual placement, but in the final design a strap will ensure constant contact. Secondly, the position of the sensor on the wrist can have an impact on the accuracy of the reading, with most accurate readings being expected when the sensor is positioned on the ulnar artery.

### C. Blood Pressure

Blood pressure (BP) is a vital measure for assessing the haemodynamic status of a patient, and taken in combination with other measures, it informs the clinician of their overall status and aids them in formulating both a diagnosis and management plan. It is conventionally measured using a sphygmomanometer, a device that includes an inflatable cuff that sits on the upper arm and is inflated to restrict blood flow through the brachial artery. For a compact, non-intrusive and continuous BP tracking system this is not viable. Currently, the only known method for directly measuring BP relies on restricting blood flow and then slowly releasing it, however recent studies have determined various methods that demonstrate how blood pressure can be inferred from other measurable parameters. These distinct methods can be combined to provide a higher accuracy BP measurement. Below three methods are outlined, which will be included in the device to allow BP to be inferred.

*Pulse Transit Time (PPT):* Pulse transit time refers to the time it takes for a pulse wave to transmit through the body. Using an electrocardiogram (ECG) device to measure a precise reference for the hearts rhythm, and a standard heart rate monitor (which measures pulse through any non-electrical method) to observe when the pulse wave travels past a particular point in the body, a time value which is proportional to the pulse transit time can be determined. By then applying the Moens–Korteweg equation a value for the arterial stiffness can be calculated, and through a series of relationships defined by Ruiping Wang et Al [9] BP can be estimated.

*Photoplethysmogram Intensity Ratio (PIR):* The PIR refers to the ratio between then peak intensity $I_h$ and the valley intensity $I_L$ (shown in Figure 1, see Appendix C) of a PPG waveform. The PPG waveform is a plot of the intensity of light absorbed by a PPG sensor when light is emitted from a source in contact with the skin. As the BP changes throughout a pulse cycle, the arterial width varies, which affects the level of light absorption of the body and hence is shown in the PPG




(b)

| Parameters of Pulse Wave | Definitions |
|---|---|
| $S_{amp}$ | Systolic peak amplitude |
| $R_{amp}$ | Reflective peak amplitude |
| $N_{amp}$ | Notch peak amplitude |
| $S_{time}$ | Systolic peak time |
| $R_{time}$ | Reflective peak time |
| $N_{time}$ | Notch amplitude time |
| $P_{time}$ | Period time |

$A_s$ : the area of systolic
$A_d$ : the area of diastolic

Fig. 2. Typical magnetoplethysmogram waveform and relevant points

plot. Using relationships defined by Xiaorong Ding et al. [5] and the value of PTT measured through the method above, a estimate of BP can be made.

*Magnetoplethysmogram Pulsimeter (MPG):* A magnetoplethysmogram consists of magnets placed over the radial artery and corresponding Hall Effect Sensors measuring the displacement of the magnets due to the increase in volume of the artery with each pulse as a voltage. A typical resulting waveform is seen in Figure 2. The systolic and diastolic blood pressure can then be calculated using the following formulae:

$$P_s \text{ [mmHg]} = \frac{1 + \dfrac{N_{amp}}{S_{amp}}}{\dfrac{S_{amp}}{N_{amp}}} A_s \tag{5}$$

$$P_d \text{ [mmHg]} = \frac{1 + \dfrac{N_{amp}}{S_{amp}}}{\dfrac{S_{amp}}{N_{amp}}} A_d \tag{6}$$

All figures and formulae courtesy of Lee et al. [11].

*1) Design:* BP is being measured in two different ways, firstly by calculating PTT from the combination of PIR and ECG, and secondly using the MPG. The ECG was constructed using three circuits: a differential amplifier with a gain of approximately 100; a notch filter to remove 50Hz frequency in order eliminate any UK mains noise and a low-pass filter with a 3dB cut-off of 130Hz to eliminate higher frequency noise. Circuit and Bode diagrams can be found in Appendix C. The prototype ECG can be seen in Figure 3. Electrodes

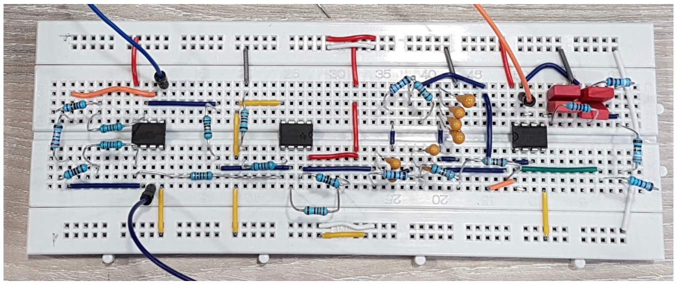


Fig. 3. 3-lead ECG breadboard

were then connected to a test subject and input to the circuit to successfully display an ECG plot on an oscilloscope. This demonstrated that a sharp peak (denoting the QRS complex of the cardiac cycle - see C, Figure 14) can be produced This can be used as a precise reference point for the cardiac cycle and will be used to calculate PTT. The test subject's ECG plot can be seen in Appendix C, Figure 15. Currently, the ECG requires 3 electrodes - one connected to each wrist and one ground electrode connected to the test subjects leg. In the final product a 2-electrode ECG would be required, as it is not practical to have an electrode connected to ones leg, however within the time-frame of this project it is more important to demonstrate the estimation of blood pressure from the combination ECG and PPG. If time remains at the end, designs of 2-electrode ECGs will be investigated.

## III. Costing Table

TABLE I
COMPONENTS COST

| Component | Quantity | Price (£) | Total (£) |
|---|---|---|---|
| Thermistor NTC | 2 | 0.679 | 1.358 |
| Sensirion STS35-DIS | 1 | 2.34 | 2.34 |
| Temperature lm75 | 2 | 0.443 | 0.886 |
| Heart Rate MAX30100 | 1 | 18.88 | 18.88 |
| Arduino Uno Click Shield | 1 | 6.03 | 6.03 |
| Double Sided Velcro Strap | 1 | 5.99 | 5.99 |
| Velcro Dots | 1 | 2.4 | 2.4 |
| Bag | 1 | 3.99 | 3.99 |
| Miniature Hall Effect Sensor | 5 | 2.24 | 11.2 |
| N42 Neodymium Magnet | 1 | 4.95 | 4.95 |
| Wire | 10 | 0 | 0 |
| Battery Pack (In stock) | 1 | 0 | 0 |
| USB Cable (In stock) | 1 | 0 | 0 |
| **Total Cost** | | | 58.024 |

## IV. Experiments

### A. Temperature Sensor - Beta Value Characterisation

An experiment was carried out to find out the beta values of the two NTC thermistors for the forearm and environmental temperature readings. This would allow temperature readings to be found directly from each thermistor resistance, by the beta parameter equation 1. Although the manufacturer of the NTC thermistors gives a beta value of 4300 , [8], they also give a tolerance of $\pm 3\%$ and hence it useful to determine the exact beta values in order to obtain accurate readings.

*1) Experimental Setup and Procedure:* Each NTC thermistor was placed in a water-tight plastic bag in a cup of warm water, along with a thermometer. They were each connected to a multimeter set to measure their resistance at different temperatures in the range of interest (between 45°C and 20°C).

*2) Results and discussion:* The beta equation (1) was rearranged to:

$$\beta = \frac{ln(R_{T2}) - ln(R_{T1})}{\frac{1}{T_2} - \frac{1}{T_1}} \tag{7}$$

Where $T_1$ and $T_2$ are any two different temperatures at which the resistance $R$ is measured. From this relationship it was found that the beta value could be determined from the gradient of a plot of $ln(R)$ against $\frac{1}{T}$. A graph depicting these results, Figure 5 and 6, can be found in Appendix A. They indicate different beta values for each thermistor, 4107 for thermistor 1 and 4119 for thermistor 2. These results are very accurate with $R^2$ values of 0.9997 for both thermistors. The lower bound indicated on the data sheet is 4171 so these values are lower than expected. As the beta values were calculated using the same leads planned to be used in the final device it was decided to use these values rather than the lower bound value from the data-sheet. Initial testing supported this decision providing more accurate results than the data sheet beta values when compared to temperatures recorded by the digital thermometer.

### B. Temperature sensor - finding accuracy of sensor

*1) Experimental Setup and Procedure:* Data was collected for the skin and environment temperatures over ten minutes, and was compared with readings taken by an IR thermometer at one minute intervals. The code also ran the conversion to core temperature, using equation 2, and this was also plotted. The graph is shown in shown in Appendix A, Figure 7.

*2) Results and Discussion:* The temperature sensors appear to be fairly accurate, with a maximum temperature difference of 0.8°C found for the skin temperature sensor, and 1.0°C for the environment temperature. When combined in equation 2, a fairly stable and reasonable core temperature for the healthy participant was predicted. Equation 2 captures the fact that whilst environmental and skin temperature may vary a lot, the core temperature will remain fairly constant until large fluctuations in skin or environment temperatures are observed.

### C. Blood Pressure

Currently no testing to calculate blood pressure has been performed, however given recent completion of the PPG sensor we hope to over the next few days see first results for PTT.

### D. Heart Rate

To investigate the accuracy of the heart rate and pulse oximeter, the Arduino set up was built using the MAX30100 sensor. The device provided oxygen readings, in agreement with normal oxygen levels expected in the human body. The exact accuracy of the readings is to be determined in comparison to medical devices. Variations in the heart rate were strongly correlated to wrist movement. Therefore, in further iterations an averaging algorithm is to be implemented such that the readings over multiple samples are considered, complemented by a decision code that rejects as invalid samples that exceed by far the average value.

## V. Progress Made and Further Work

### A. Temperature Sensor

Two temperature sensor circuits have been made on a breadboard and code has been developed which when uploaded to the Arduino is able to predict a measurement for core body temperature when one thermistor is used to measure skin temperature and one is used to measure environmental temperature. LEDs have been implemented for debugging purposes. Experiments have been performed to determine each thermistor's beta value in order to calibrate the thermistors. The devices now must be soldered onto PCB and integrated with the Velcro strap ensuring the skin temperature sensor is adequately insulated, and the code must be developed to include indication of when the temperature is abnormal.

### B. Heart Rate and Blood Oxygen Sensor

After the first week, the device is able to detect heart beats, and measure blood oxygen concentration. Without complex equipment, we cannot accurately determine the blood oxygen concentration. However, it is returning very reliable readings that are within the accepted normal range of 95-100%. Over the coming weeks, the aim is to further test the accuracy of the heart rate detection by comparing values to medical devices, alter the code such that it can exclude anomalies and take averages, input parameters that specify healthy conditions, and attempt to integrate the sensor with the ECG to calculate blood pressure.

### C. Blood Pressure

Initial work consisted of investigating as many ways of measuring blood pressure as possible and analysing their viability for this project. Not knowing which one would work best, all three of PTT, PIR, and MPG where settled on in order to be able to compare their agreement and accuracy in testing. With PPG being developed by the heart rate and blood oxygen sub-team, what remained to be done for PTT and PIR was to develop and ECG sensor. This took up most of the second week, however successful results where achieved. With ECG completed, attention was turned to the MPG. Current work involves attempting to read the output voltage of a single Hall Effect Sensor from an Arduino. Once that is achieved, the physical casing for the magnets and hall effect sensors will need to be designed, and finally the MPG waveform will need to be analysed to give blood pressure readings.

## VI. Social Aspects

### A. Summarize the Results and Discuss the Social Impact

There are many communities across the world that live in hard to reach, remote areas. They often require treks through different terrains to access or alternatively expensive and disruptive modern transport such as helicopters. For example, in Sub-Saharan Africa 29% of the population lives over 2hours away from the nearest hospital by road. [10] This makes it hard to monitor their individual and collective health. This wearable device will enable healthcare professionals to monitor their health from urban areas rather than having to continually access the area, incurring additional costs and inconvenience.

Through monitoring their vital signs alerts will be given if individuals are going through a healthcare emergency and more critically if a healthcare epidemic is erupting. Critical values of vital signs in multiple community members could indicate more than natural illness. This would often go unnoticed until it was too late in rural areas. However, early intervention can reduce the spread of the illness and allow medical staff to attend to the community rapidly.

### B. How would you Mass Manufacture?

For the initial prototype an Arduino Uno was used, mainly due to familiarity and design flexibility. In mass manufacture a custom, much smaller microprocessor would be used with only the functions necessary to this device in particular, reducing the size and cost. The electrical components would require tighter tolerances for their parameters, such as in the case of beta value of the NTC thermistors, as the additional step of calibration is costly in terms of mass manufacture. The final circuits would be constructed on a purpose-built PCB board where the size of components and electrical pathways can be significantly reduced in size, and the construction can be automated to reduce manufacturing times on a mass scale.

### C. Summarize the Economics and Logistics

The wearable healthcare device designed aims to provide people in remote areas with means of monitoring their vital signs. A fundamental requirement for the successful deployment of this technology is the small size and affordable costs.

The creation of a working prototype costed £58.024 in components. This includes components that were used for testing and comparison only. Therefore, the estimated cost of a device with a similar functionality is at £32.72. Cost optimisation can be performed and smaller and cheaper components can be used (for the MAX30100 sensor), therefore reducing the overall price.

In terms of the logistics, the device would be most suitably deployed in bulk to reach local communities. As far as packaging and transportation is concerned, the focus would be on ensuring a suitable temperature that maintains the accuracy of calibrated electrical components.

## VII. Future Scope

### A. Problems and Solutions

With a time constraint and lack of facilities available to provide tightly controlled thermal climates within the department, out team relied on the data collected by *Webb, 1992* to determine equation 2 used for the temperature sensor, however this experiment was only conducted on six men. If we were to take this project further, we would hope to collect data on several vital measurements on a much larger group of people, with a range of sexes and ages, such that a more accurate regression analysis could be performed, perhaps calibrated differently for each sex or age range.

Another problem is the fact our final prototype will not be as 'wearable' as other watch-style health trackers available on the market, due to time constraints. Thinking beyond the scope of this course, the suggested size reductions in section VI-B would inevitably make the device more wearable, as well as the implementation of smaller, more efficient batteries. Ideally these changes would enable us to fit the whole device inside a watch-style casing, and time could be spent designing this casing to be sweat-proof and more protective of the components than a Velcro strap, perhaps made out of a silicon material.

### B. Who will take this forward

Although our group is not affiliated with a specific company, we have been in direct communication with some medical students who have expressed interest in this device. Depending on the success of the remaining project, our team is considering the possibility of taking our initial prototype beyond the scope of the project.

## VIII. Conclusion

In the duration of two weeks, through structured research, planning and testing, this team has developed designs for a temperature sensor, a heart rate and blood oxygen sensor, and a blood pressure sensor, with the view of creating and integrating these devices with an Arduino Uno in a single wearable device on the user's wrist. Progress has been made on all three sensors, and the results indicate that we are able to monitor heart rate and temperature successfully.

## IX. Individual Contributions

### A. Alistair Goodman

My time has been spent working along side Daniel. Initially we investigated the background of blood pressure, how it is clinically measured and searched for different possible methods for cuff-less measurements, then we decided which of the methods would be plausible for our device and ordered the necessary components. The second half of our time was used successfully building our own ECG device, which has delivered accurate readings. Over the remainder of the project I will investigate reducing the ECG in size, integrating it with the PPG sensor to allow for BP measurements to be made and then attempting to produce the Magnetoplethysmogram Pulsimeter device.

### B. Ana-Maria Marcu

My initial task was to create a components list required for order. Once the components have arrived I have performed the required soldering to assemble the Arduino Uno board with the Click board and MAX30100 sensor. The rest of the first week was spent working together with Joe on the heart rate sensor, performing the debugging and implementing a mean of signalling when heart beats happen. I now aim to identify healthy and unhealthy patterns in heart rate and implement an algorithm that is able to automatically detect them from the recorded data, as well as supporting the integration with the other sensors.

### C. Daniel Jackson

I have worked closely with Alistair in developing a blood pressure monitor. The first week was spent researching different methods of measuring blood pressure before settling on cuff-less measurement. We then developed a functional prototype ECG. I will now make a start developing an MPG, which requires the design and manufacture of a silicon rubber housing for the magnets, followed by the analysis of the waveforms produced.

### D. Joe Story

My first task was to source the components needed for the PPG pulse detection. After the components arrived, the rest of the first week was spent with Ana-Maria, implementing the MAX30100 heart rate and pulse oximeter sensor with the Arduino Uno. This involved researching blood oxygen content calculations (Sp02), researching relevant Arduino libraries needed to communicate with the sensor, and developing the relevant code. Now that the board is able to return accurate blood oxygen and heart rate readings, my next goal is to improve the code further so that it can average out the readings, and exclude anomalous results. I also aim to assist the multiple sub-teams with Arduino integration as this is the area of experience that I believe I can bring to the team.

### E. Rose Humphry

My time was spent primarily developing the temperature sensor. I spent some time researching into the different types of temperature sensor, as well as the relationship between skin temperature and core body temperature. I then helped design the NTC thermistor circuit and helped test and develop the code that translated its resistance into a temperature reading. I was also involved in the experiment to find the beta values of our thermistors. I hope to now focus on refining the temperature sensor into a more wearable, and helping to assist with the integration of the sensors.

### F. Vidya Kanakaratnam

Initially, I spent my time identifying key potential problems we may encounter and developing the problem statement. I then moved on to assist in developing the temperature sensor. This involved researching and developing the NTC thermistor into an accurate temperature sensor. This has been predominantly through writing the code to link the temperature sensor readings from the skin and micro-climate to the internal core temperature. Various relationships found in literature were tested to identify a good preliminary result. In addition, I participated in experiments to identify the Beta values of the NTC thermistors being used. For the remaining time I will help fine tune the temperature sensor and help integrate the components along with the rest of the team to get a complete device.

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# Appendix A
# Temperature Sensor

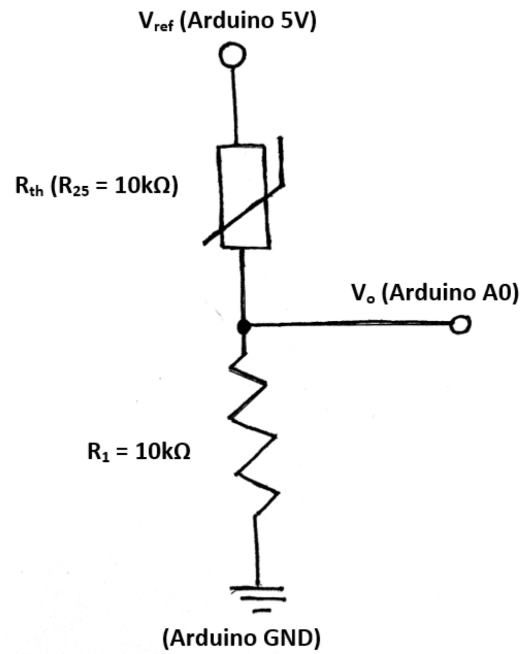


Fig. 4. NTC thermistor circuit diagram. $R_th$ is the NTC thermistor resistance, $R_1$ is the resistance of the resistor in the voltage divider, $V_o$ is the voltage output, connected to the analogue pin (A0) of the Arduino's ADC, $Vref$ is the input voltage, connected to the 5V supply from the Arduino.

TABLE II
Data obtained from *Webb, 1992* ( [7]), measuring the left forearm skin temperature ($T_{fs}$) and the auditory canal temperature ($T_c$) of six different men at different environmental conditions ($T_e$).

| $T_e$ (°C) | $T_c$ (°C) | $T_{fs}$ (°C) |
|---|---|---|
| 15.0 | 35.9 | 26.9 |
| 27.0 | 37.0 | 34.0 |
| 47.0 | 37.8 | 37.0 |

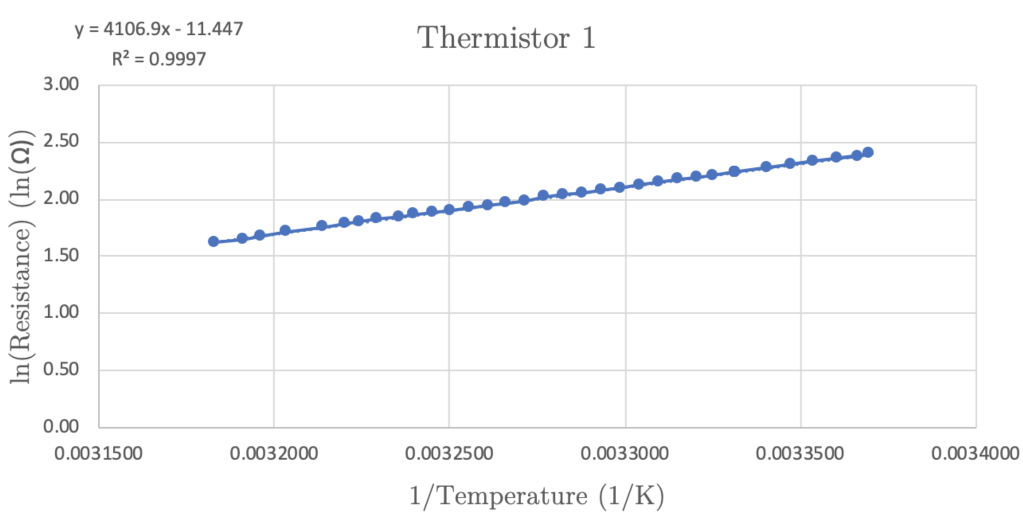


Fig. 5. Plot of the natural logarithm of Resistance against the inverse of Temperature for Thermistor 1

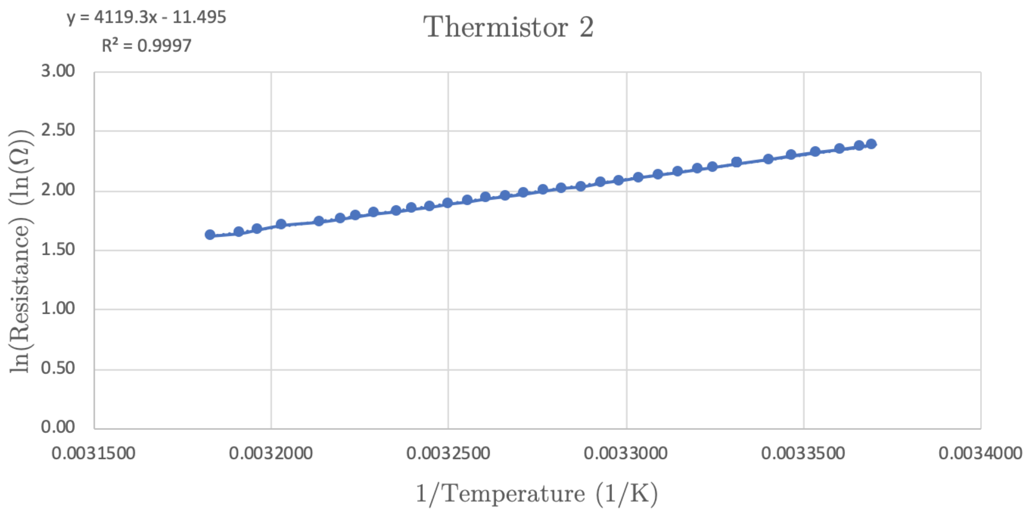


Fig. 6. Plot of the natural logarithm of Resistance against the inverse of Temperature for Thermistor 2

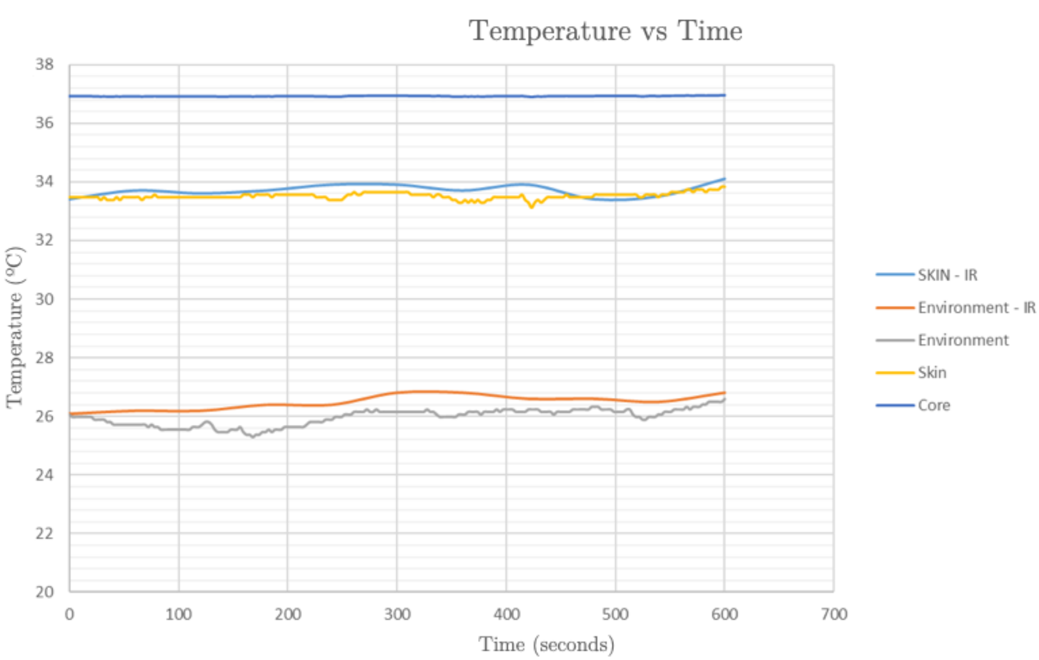


Fig. 7. Plot of Temperature against Time for the temperature of the the environment, insulated skin and core temperature. "Skin IR" and "Environment IR" are measurements direct from the infrared thermometer. "Environment" and "Skin" are readings from the NTC thermistor calibated by the beta parameter equation and "Core" is calculated from these two readings as representative of the auditory canal temperature.

# APPENDIX B
# HEART RATE

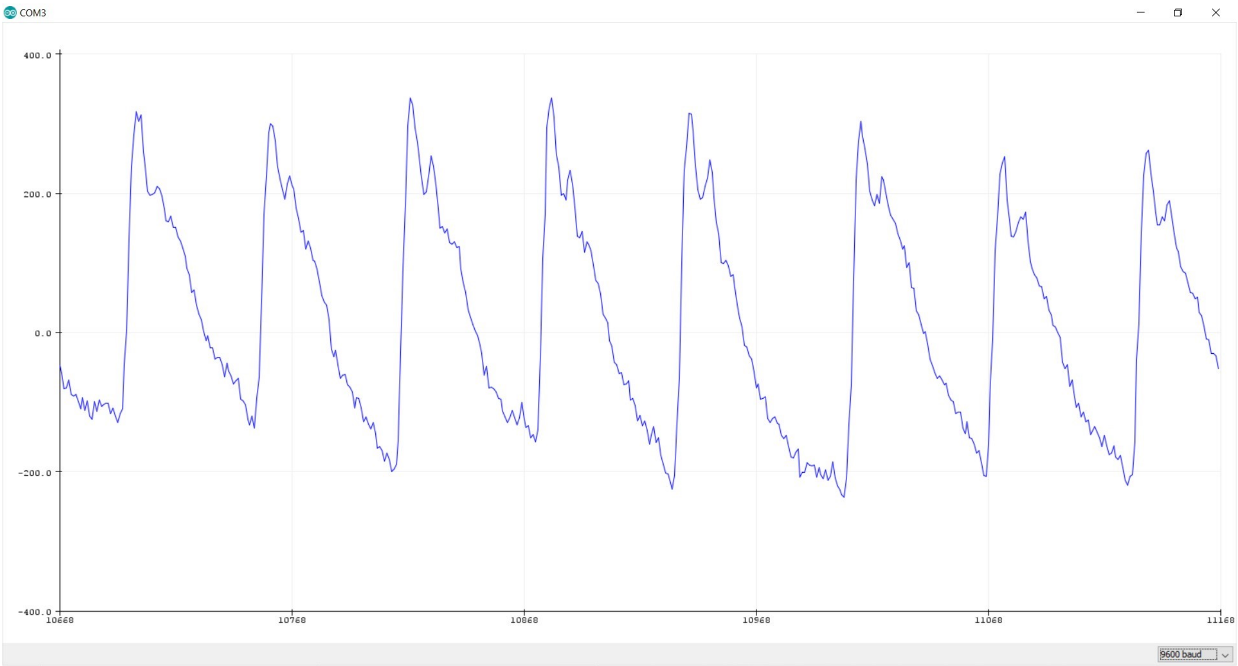


Fig. 8. Arduino plot of the averaged infrared and red detected DC values to show the pulsatile components of heart rate

# APPENDIX C
# BLOOD PRESSURE

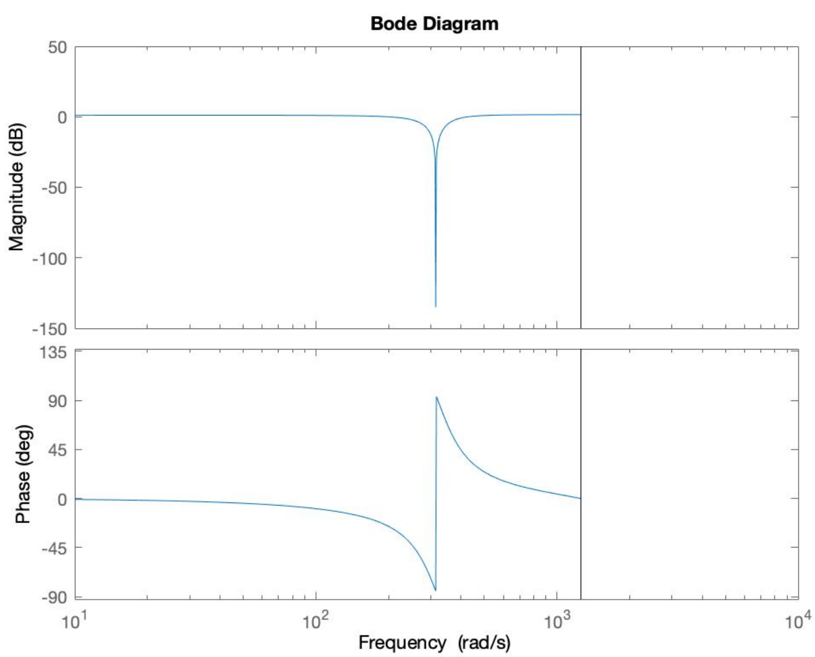


Fig. 9. Bode Diagram of Notch filter

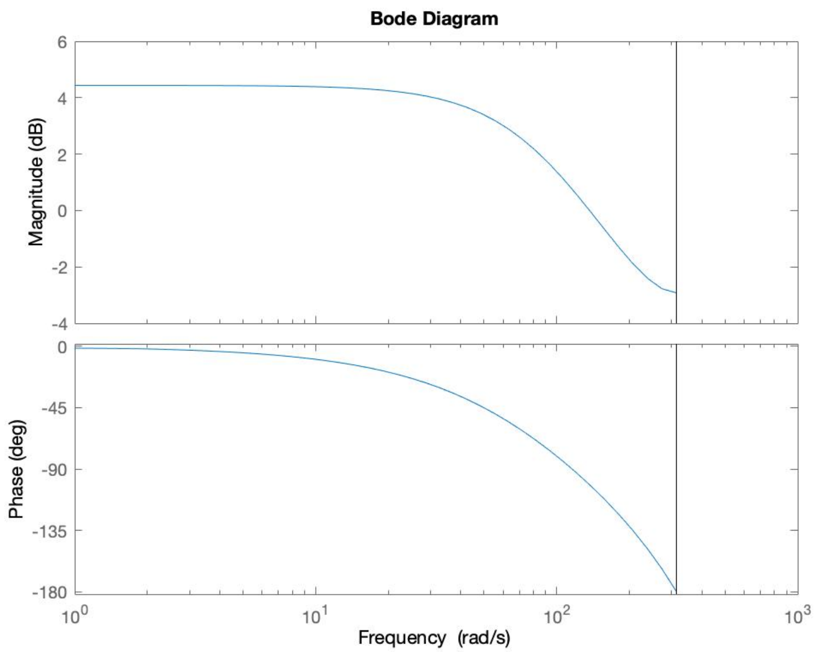


Fig. 10. Bode Diagram of low-pass filter

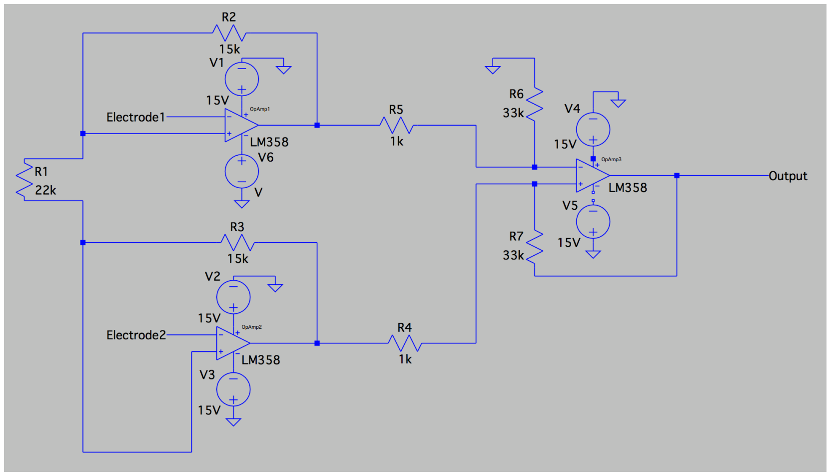


Fig. 11. Differential amplifier circuit diagram

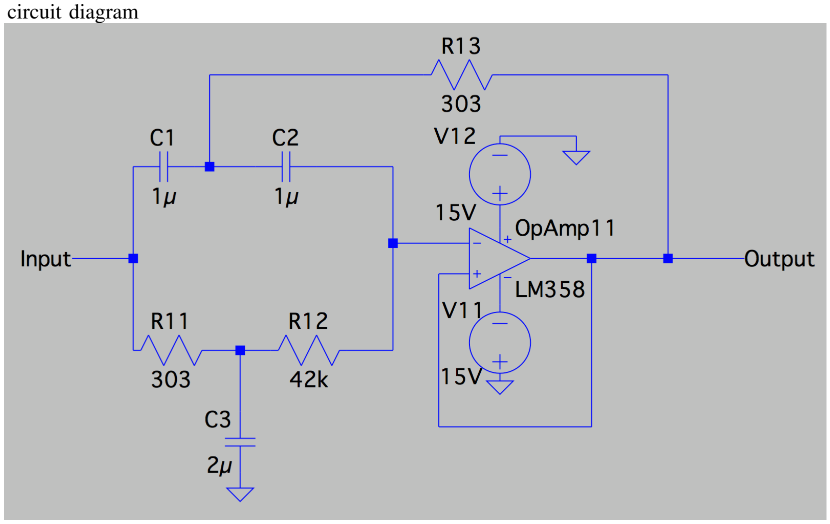


Fig. 12. Notch filter circuit diagram

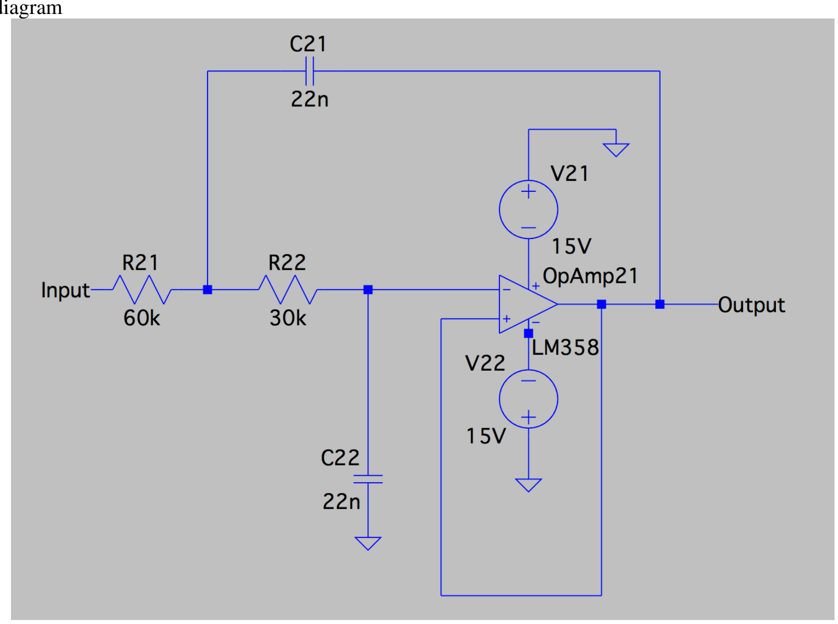


Fig. 13. Butterworth low-pass filter

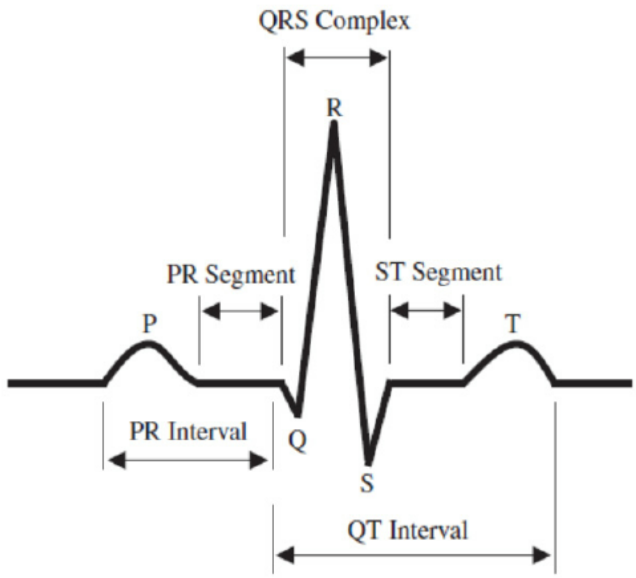


Fig. 14. Typical ECG waveform

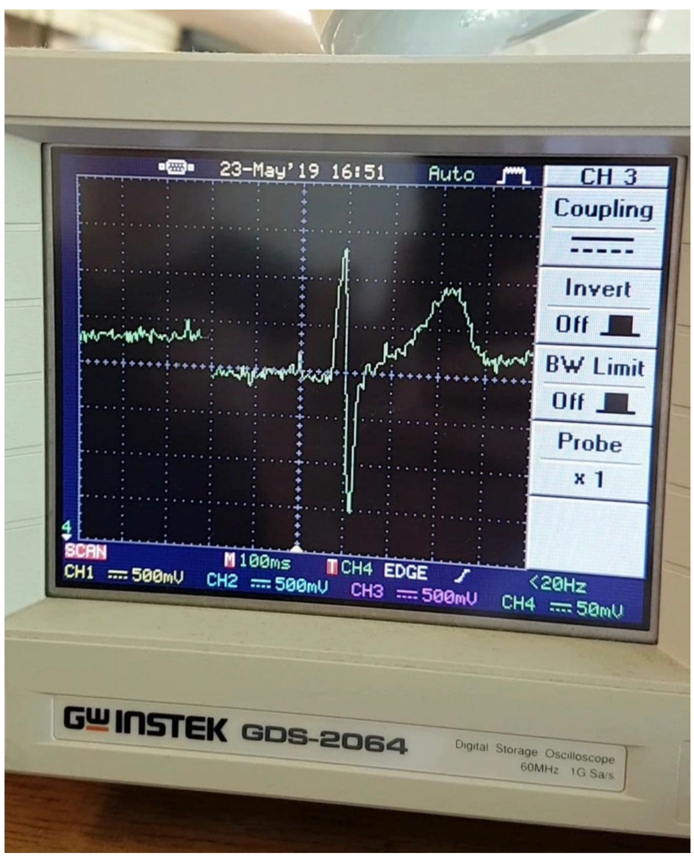


Fig. 15. Test subject's ECG