Temperature based heart rate detection
Heartbeat measurement in a wireless headset

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TEMPERATURE BASED HEART RATE DETECTION

HEARTBEAT MEASUREMENT IN A WIRELESS HEADSET

by

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This thesis, assigned by Plantronics®, contains the implementation of a temperature based heart rate sensor as an alternative to established sensing methods. This is realised by the use of 0402 package NTC thermistors in a bridge configuration. The data acquisition is done by the use of a National Instruments USB-6210 data acquisition module. This data acquisition module sends information to a computer, where it is processed in MATLAB. The acquired data is then filtered and the heart rate is determined from the frequency spectrum. In the end, the system was able to measure the heartbeat on several locations on the human body and temperature differences up to 0.05mK.
The project started on the 20th of April 2015 and will be concluded on the 3rd of July, totalling just short of eleven weeks.

A proposal was done by Dr. ing. G. de Graaf in association with Plantronics®. The project team, consisting of six members, explored this proposal. This proposal consisted of creating a heart rate sensing module that can be interfaced with a wireless headset. However, after a discussion with Plantronics®, this proposal was modified into a research project in which different techniques for measuring the heart rate had to be researched. During this project, a lot of technical experience was gained, as well as project management skills.

Our thanks goes out to Dr. ing. Ger de Graaf and Ing. Ron van Puffelen for supervising our plans and progress, providing us with critical hardware, and advising us. Furthermore we would like to thank Plantronics® for the opportunity to work on this project. Special thanks goes out to our fellow group members: J. Guyomard, J.A.G. Jonkman, T.M. de Rijk and R. Stortelder. They provided help, insight and extra motivation to complete this project.

D.C. Kaandorp & A.I. Kanhai
Delft, June 2015
In this age of gadgets and smart devices, manufacturers try to add as many features to a device as possible. A well known example is a smartphone, but smartwatches are also on the rise, as are many other devices like smart glasses. Another trend is seen in a entirely different sector: sports. With all the new smart devices and their ability to track via GPS, a rise in sporting activity can be observed. While some people are competitive with their friends, others just want to be able to check their progress or post their results on social media. There is one element that lacks in most smart sport devices: audio. Of course it would be anti-social to use the smart devices' speakers and let the public enjoy your music, so the common solution is to use earphones or a headset. To solve the complication of wires around the head, wireless headsets were invented, such as the BackBeat FIT[1] by Plantronics® [2], which can be seen in Appendix B. To combine the sports and new smart gadgets, Plantronics® opted to make an addition to the BackBeat FIT. Not only is it able to stream music from one's smart device and able to accept calls, it will also be capable to measure heart rate and display this on the smart device. It will also be possible to get live audio feedback by pressing a button on the headset.

Together with four other teammates, research was done regarding different methods of heartbeat detection, keeping in mind that the measuring location is in or around the ear. Plantronics® suggested ideas for researching alternative sensing techniques that might react differently to the limitations of current heart rate modules. From our literature study and discussion with Plantronics®, three different methods arose:

- Photoplethysmography (PPG)
- Temperature based measurement
- Skin discolouration

For this thesis, the feasibility of the second method was researched. In the next chapter, the ethical problems concerning our system will be discussed. Then the system requirements will be mentioned, followed by the relevant research and theory regarding this topic. After this, the design phase will be discussed, followed by the results. Lastly these results will be discussed.
This chapter discusses the ethical problems that can occur regarding the heart rate sensor.

The primary application for the sensor is integration into the BACKBEAT FIT\([1]\). Since this is a sporting headset, deviations in the displayed heart rate do not have major consequences. They are tolerable, because one does not care for the exact heart rate. However, this can induce problems when the system is used for other purposes, e.g. medical purposes. When measuring heart rate in a medical setting, it is extremely important that the required data is as valid as possible. If not, it can lead to disastrous consequences. Our system has a limited bandwidth that only includes relatively safe heart rate regions. Going under 40BPM or above 240BPM will result in false or no output. Thrusting the given output while outside the bandwidth might lead to very dangerous situations. Our module is only meant for recreational use and can not be accounted for in the case medical problems due to incorrect heart rate feedback.

Another ethical problem that can occur is abuse of the data by insurance companies. From one’s heart rate a lot of medical data can be derived. A trained cardiologists can derive signs of a number of heart diseases, such as ventricular hypertrophy and chronic lung disease[3], from a medical ECG test. Because a temperature derived heartbeat signal contains a lot less information, deriving medical data is more limited, but still a lot of information that would be of interest to insurance companies can be derived from this system. The heart rate module does not have direct influence over how the data is used, but one could consider only supplying the module to companies with strict ethical policies.
The system should satisfy a number of requirements to be called a success, which are listed below.

3.1 The system should be able to measure the heart rate on the human body, preferably inside the ear.

3.2 The system has to operate in 0.67-4Hz or 40-240BPM, the minimum and maximum human heart rate[4].

3.3 The system should have a low power consumption compared to ~40mW of bluetooth[5].

3.4 The sensing end of the system has to be small enough to fit inside a earphone.

3.5 The system has to be affordable for consumer electronics.

Requirements 3.1 and 3.2 are needed to create a successful heart rate sensing module. Requirements 3.3 - 3.5 are important in order to, during further development (out of the scope of this thesis), integrate the module inside a wireless headset.
RELEVANT RESEARCH

Nowadays, the most used techniques to measure heart rate are photoplethysmography (PPG) and electrocardiography (ECG). These techniques are however heavily influenced by motion artefacts during physical exercise. The reduction of these motion artefacts on PPG is researched by an other subgroup of this project [6]. They will try to reduce these motion artefacts by the use of an accelerometer. Another subgroup will research a technique using skin discolouration to determine the heart rate [7]. This change in colour of the skin is caused by the change of blood flow behind it.

High resolution temperature measurement to determine heart rate is not a highly researched topic, which is the reason Plantronics® suggested this subject. However, studies have shown that it is possible to obtain the heart rate using this technique [8]. This was done on locations were arteries were close to the surface, like the neck and wrist. Since the sensor should have the possibility to be integrated in a headset, the best location to measure around the ear was researched [9][10][11]. The most important aspect of good location when using a temperature sensor is the presence of a major artery. It was concluded that the best location around the ear would be right in front of the ear on the superior temporal artery [10]. However, this is not practical for implementation in the BackBeat FIT, visible in Appendix B, since this integration will have to alter the original design. Also, it is not comfortable for the user and will decrease the ease of use. Apart from these reasons, placing the sensor outside of the ear might be detrimental for the performance of the system. This is because outside the ear the sensor is highly subjected to external thermal changes. Due to these reasons, it was decided to research the performance of the sensor inside the ear, since this provides a better thermal isolation from the environment.
This chapter discusses the relevant topics before the design and implementation phase. This consists of the main principle that is used, the possible sensor types, and possible problems that have to be evaluated during the design phase.

5.1. **Principle**
The reason it would be possible to measure heart rate through the use of temperature is because the core of the human body is, in most climates, hotter than the extended parts. This temperature difference will result in cooling of the blood when outside of the core of the body. The blood pressure in the arteries changes with every heartbeat, which will mean that the blood will not flow with a constant speed. This results in some of the blood having a longer cooling down time, thus creating temperature difference. To measure the heart rate by means of temperature fluctuations it is necessary to be able to measure temperature differences up to 0.2mK[12].

5.2. **Temperature sensor**
Two common types of temperature sensors were considered, of which one was implemented.

5.2.1. **Infrared Detectors**
IR technology is a widely used technique for temperature measurement and control. IR technology is based on the fact that an object emits thermal radiation, also known as black-body radiation, which can be measured. The optic resolution of an IR sensor is related to the following ratio:

\[
DS_{\text{ratio}} = \frac{d}{A}
\]

(5.1)

where \(d\) is the distance from the sensor to object and \(A\) is the area from which the radiation is received and measured. In our case, the preferred situation is when motion of the sensor relative to the ear has minimum influence on the output. This can be realised by averaging over a relatively large surface. When placing the sensor inside the ear, the measurable surface with a IR-photo-based sensor is very small because of the small distance. This will in turn mean that movement of the module can point the sensor at an entirely different surface, resulting in undesirable signals. That is the main reason why it was decided to opt with a more rigid sensing system, which will be discussed next.

5.2.2. **NTC thermistor**
Another common type of temperature measurement is by the use of an NTC thermistor. An NTC thermistor responds to physical changes in temperature by changes in its resistance. The resistance of an NTC thermistor is related to the temperature by the following equation[13]:

\[
R_T = R_0 e^{\beta \left(\frac{1}{T} - \frac{1}{T_0}\right)}
\]

(5.2)
where $R_0$ is the resistance at $T_0 = 298.15K$, $T$ is the temperature, and $\beta$ is a constant that describes the sensitivity of the thermistor.

An important feature of an NTC thermistor used for time dependent signals is the response time. The response time has to be sufficient to be able to react to the frequency range of the heart rate specified in Chapter 3. The response time is mainly dependent on the thermal capacity of the sensor, meaning that a smaller package would result in a smaller response time[14]. Because a thermistor is influenced by the surrounding temperature, it is less subjective to motion. Besides, it is also extremely easy to implement in comparison with a IR sensor. This was the decisive factor to continue with a NTC thermistor.

### 5.3. Noise

Since the change of the resistance of the thermistor (and thus the voltage) due to temperature fluctuations of the body is very small, noise has to be addressed properly.

There are a few noise sources present in the system. The voltage source that will supply the system contains noise. The reduction of this noise will be addressed in section 6.1. The resistors also have thermal noise, which is given by:

$$V_T = \sqrt{4kTBR}$$

(5.3)

where $k = 1.38 \cdot 10^{-23} \text{JK}^{-1}$ is the Boltzmann constant, $T$ is the temperature, $B$ is the bandwidth, and $R$ is the resistance. However, by choosing an appropriate value for the resistors, this noise is negligible.

Other components like the operational amplifiers and instrumentation amplifiers have an input voltage noise and input source noise. Because the signal exists in the low frequency range, the amplifier input noise has 1/f, or flicker noise[15]. This means that there is a possibility that the signal has to be modulated in order to shift the signal into a higher frequency region to reduce this noise.

Other noise includes primarily thermal noise from the air, since the temperature can fluctuate depending on the environment. This requires adequate isolation of the thermistor from the air.
This chapter describes the design and implementation phase. This also includes the data acquisition in MATLAB, and a brief verification of the first results after signal processing.

6.1. **Bridge-Circuit**

A bridge-circuit will be used to measure the temperature fluctuations. This design was chosen because a differential amplifier can be placed after the bridge. Due to this configuration, the two correlated signals originating from the source can be subtracted. This will result in a reduction of the amplitude of the noise and the DC-offset at the output of the bridge. Two type of bridges will be discussed, displayed in Figure 6.1. The circuit in Figure 6.1a was the initial design. This bridge consists of four NTC thermistors. Two of these are thermally isolated thermistors (diagonally across). This will cause them to respond slower and less to temperature variations than the other two thermistors, this principle is displayed in Figure 6.2b. By creating this reference signal of the absolute temperature, the resulting output has a lower DC-component. The ideal case is when the offset is completely filtered after the bridge, which is not realisable. This idea is displayed in Figure 6.2a.

However, isolating the thermistors sufficiently turned out to be harder than expected during prototyping. This was mainly due to the size, and therefore it was decided to continue with the circuit in Figure 6.1b. This circuit lacks the DC reduction explained above, and thus needs additional filtering after the bridge.

![Figure 6.1: Two types of bridge circuits](image-url)
Using Equation 5.2, the resistance change can be calculated: \[ \Delta R = -7.0275 \cdot 10^{-6} \], with \( \Delta T = 0.2 mK \), \( T = 310.15 K \), and using \( \beta = 3380 \) as a starting point (based on availability). Whether this is feasible will be determined by the noise. Filling in all parameters except \( R \) in Equation 5.3 leads to \( (2.4565 \cdot 10^{-10}) \cdot R \). An appropriate value has to be chosen to make this thermal noise negligible. By choosing a factor of \( \frac{1}{100} \) \( \Delta R \) w.r.t. thermal noise, a resistor value can be chosen. This results in 10k\( \Omega \), which is tolerable. Using this information and the required package size resulted in choosing the NCP15XH103J03RC[16].

### 6.2. AMPLIFICATION AND FILTERING

Behind the bridge circuit, two high-pass filters were implemented. The cut-off frequency of these filters are: \( f = \frac{1}{2 \pi RC} = 0.1592 Hz \). The component values were chosen based on requirement 3.2 and availability. After filtering, the signal has to be amplified. For amplification, the INA126PA[17] was used. This is a low noise instrumentation amplifier that was already available in the lab. It has an input noise voltage of 35nV/\( \sqrt{Hz} \). However, in the range of 0.1 Hz - 10 Hz this increases to 0.7µV. This is a factor \( \frac{0.7\mu V}{35nV} \approx \frac{1}{100} \) of the signal.

The necessary gain of the instrumentation amplifier was determined with the assumption that the temperature fluctuation would be around 0.2mK - 1mK[12]. It needs to be high enough to display the heartbeat, but not too high that the absolute temperature fluctuations, caused by the environment, would result in clipping. An appropriate amplification gain for this requirement turned out to be 805. At the output of this amplifier there was a DC component present in the signal. To further reduce this, a capacitor was placed after the INA126PA. The resulting circuit is displayed in Figure 6.3.

A relation between the voltage and temperature can be derived. Using \( T = 310.15 K \), \( \Delta T = 1 mK \), \( T_0 = 298.15 K \), \( R_0 = 10k\Omega \), \( \beta = 3380 K \), \( G = 805 \) and \( V_+ = 6V \) in Equation 5.2 leads to:

\[
R_{T+\Delta T} = R_0 \cdot e^{\beta (\frac{1}{T+\Delta T} - \frac{1}{T})}
\]

which is used to calculate the voltage fluctuation. This results in:

\[
\Delta V = 2 \cdot G \cdot V_+ \cdot \left( \frac{10k\Omega}{10k\Omega + R_T} - \frac{10k\Omega}{10k\Omega + R_{T+\Delta T}} \right) = 81 mV \tag{6.1}
\]

due to a 1mK temperature fluctuation. This ratio will be used during processing for easy evaluation of the temperature measured by the system. For the exact temperature the battery voltage has to be taken into account. Notice the factor 2 in Equation 6.1, this is caused by diagonally placed thermistors.
6.2. Amplification and Filtering

6.2.1. First Measurement

With the implementation of Figure 6.3 it was possible to measure heartbeat at the carotid artery. This was measured with the use of an oscilloscope[18]. The result is displayed in Figure 6.4. The measured heartbeat signal has an amplitude of around 100 mV and a frequency of 78 BPM, which was verified with a PPG heart rate monitor[19]. This would imply that the differential temperature is around 2.5 mK. Also observed from the graph is a 50Hz signal which shows the need for better shielding or a low pass filter. The changes in the absolute temperature are around 800 mV or 20 mK.

(a) Zoomed out: 200mV/div, 2s/div.  (b) Zoomed in: 100mV/div, 100ms/div.

Figure 6.4: The first successful heartbeat measurement.
6.3. SIGNAL PROCESSING

After the temperature fluctuations were displayed on an oscilloscope, the NI USB-6210[20] was used for data acquisition. This is a USB data acquisition module with build-in ADC that can sample up to 250kS/s which allows direct readout in MATLAB. The complete system is displayed in Figure 6.5.

![Figure 6.5: Overview of the complete system.](image)

6.3.1. FILTERING

For the filtering a rectangle-function was used in the frequency-domain, this removes everything outside the spectrum stated in requirement 2.4. The result of this filter can be seen in Figure 6.6.

![Figure 6.6: The signal before and after filtering](image)

6.3.2. ZERO Padding

The resolution in frequency domain is:

\[ \Delta R = \frac{f_s}{N} \]  

(6.2)

where \( f_s \) is the sample frequency and \( N \) is the number of samples. The duration of the measurements were 5 seconds, which results in a resolution of 0.2Hz or 12BPM. By the use of zero padding (making the time-domain signal a factor 12 longer), this resolution is increased to 1 BPM. This is applied in the next chapter.
This chapter contains all the measurements used to verify if the requirements specified in Chapter 3 are met, followed by the results of the in-vivo measurements (tests on human subjects). The validation is done by the use of a Peltier cooler, explained in Appendix A.

7.1. VALIDATION

In order to validate if the design meets the specifications, several tests were done. These are discussed below.

7.1.1. RESOLUTION

In order to measure the heart rate, the output noise voltage has to be low enough that a signal of 0.2mK can be measured. Using the calculated $81V/K$ in Chapter 6, results in $0.2mK \cdot \frac{81V}{K} = 16.2mV$. The Peltier setup, as stated in Appendix A, was adjusted to the point where this requirement was met. The signal used was a 0.1A, 7V, 1Hz signal. The result can be seen in figure Figure 7.1. The applied 1Hz or 60BPM signal is clearly visible.

![Figure 7.1: Peltier temperature resolution requirement validation.](image-url)
7.1.2. **BANDWIDTH**

The bandwidth is also verified with the Peltier set-up. Two measurements were done, the first at 240BPM to verify that the requirement is met. Secondly, the frequency is increased until no longer visible in the frequency spectrum. The results can be found in figures Figure 7.2 and Figure 7.3. The maximum frequency was found to be around 5Hz or 300BPM. (This does not mean that this is the maximum frequency that can be reached with the design, but rather that we cannot prove it works at a higher frequency with this set-up.)

![Figure 7.2: Bandwidth requirement validation.](image)

![Figure 7.3: Maximum bandwidth](image)

7.1.3. **POWER CONSUMPTION**

In order to validate if requirement 3.3 was met, the power consumption was measured with a precise bench-top multimeter while operating. Using $V_+ = 5.36\, V$, $V_- = -5.55\, V$, $I_+ = 0.68\, mA$ and $I_- = 0.15\, mA$ leads to:

$$P = V_+ \cdot I_+ + V_- \cdot I_- = 4.48\, mW$$

Compared with the $\sim 40\, mW$ of Bluetooth, the heart rate system in its current form will reduce battery life with about 10%. This power consumption is not optimised and there is room for improvement. For example, by reducing the supply voltage to 3.3V, around 2mW will be saved in the thermistor bridge.
7.2. IN-VIVO RESULTS

The measurements were done on several locations on the body to examine the performance of the sensor. These locations are the most common and usual places on the human body for heartbeat measurements. The locations range from most to least likely regarding heartbeat measurements. On each location, a series of measurement were done by making direct contact with the skin, and an other series of measurements were done on a distance of 1mm. This was done to see the influence of a small distance between the sensor and the skin. This distance was achieved by placing a tube on the sensor that was 1mm longer. By using a tube, the temperature noise from the environment is reduced, since the air between the sensor and the skin is isolated from surroundings. The following results were selected from series of measurements, from which later on the error rate will be discussed.

7.2.1. NECK

The sensor was placed on the carotid artery. To validate that the acquired signal is in fact a heartbeat, for the first measurement the PPG module of J. Guyomard and R. Stortelder[6] is connected to a second channel of the data acquisition module. The temperature sensor is placed against the neck while at the same time the PPG module is placed against the index finger. The results are displayed in Figure 7.4. From this image one could determine the delay between a heartbeat and the blood reaching the index finger. It is clear that the temperature signal and the PPG signal have a very strong correlation. After this validation the remaining measurements were done.

![Figure 7.4: Temperature vs PPG](image)

The results of the first measurement are visible in Figure 7.5. The FFT in Figure 7.5a shows a clear peak at 72 BPM. The same applies for the FFT in Figure 7.5b. Notice that the amplitude of the peak from the contactless measurement is actually higher than the peak from direct measurement. This may be caused by the higher cool down time the thermistors have due to the spacing. The consequence is a higher temperature fluctuation, and thus higher amplitude.
7.2.2. Toro

The sensor was placed on the skin close to the heart. The results are displayed in Figure 7.6. The most dominant peak in the FFT in Figure 7.6a is at 76 BPM. The highest peak in Figure 7.6b is at 84 BPM. Again, the peak from the indirect measurement is higher, possibly caused by the previously mentioned reason. Comparing this amplitude to the amplitude from the neck measurements, these peaks are approximately three times as small. This is expected, since at the torso the distance between skin and major arteries is greater than around the neck.
7.2.3. **Wrist**
The sensor was placed on top of an artery close to the surface of the skin on the wrist. The results are depicted in Figure 7.7. The FFT in Figure 7.7a shows an easy to detect peak at 73 BPM. Figure 7.7b does show a global maximum at 78 BPM, but has a very low amplitude compared to the previous measurements.

![Figure 7.7: The result of the measurements at the wrist](image)

7.2.4. **Index Finger**
The sensor was also placed on the index finger. The results are visible in Figure 7.8. In the FFT in Figure 7.8a the peak corresponding with the heart rate (validated with an external heart rate monitor) at 80 BPM does not have the highest amplitude. However, the highest peak is very unlikely to correspond to the heart rate, since a heartbeat signal this low only occurs at complete rest[4]. Figure 7.8b does not show this problem, and has a dominant peak at 74 BPM.

![Figure 7.8: The result of the measurements at the finger](image)
7.2.5. **EAR**

The sensor was placed inside the ear. The results are displayed in Figure 7.9. Looking at Figure 7.9a, it is visible that the FFT shows a clear peak at 71 BPM. The same applies to Figure 7.9b, with a dominant peak at 66 BPM. This location holds the most important results for our research. It has to be noted that during these measurements the sensor was inserted 14mm into the ear.

![Graphs showing results for direct contact and 1mm spacing](image)

(a) Direct contact  
(b) 1mm spacing

Figure 7.9: The result of the measurements at the ear

7.2.6. **SUCCESS RATE**

As previously mentioned, a number of measurements (up to 40, depending on the location) were done. The success rate of these measurements was determined by looking for a global maximum in the frequency domain between 65 and 85 BPM (the test subject’s heart rate at rest) with sufficient amplitude. The success rate of all the measurements are summarised in Table 7.1. The neck has the best result in both cases. It is also apparent that on all locations except the index finger, the success rate is higher when making contact.

<table>
<thead>
<tr>
<th>Location</th>
<th>Contact</th>
<th>No contact</th>
</tr>
</thead>
<tbody>
<tr>
<td>Neck</td>
<td>60.0%</td>
<td>38.5%</td>
</tr>
<tr>
<td>Torso</td>
<td>22.5%</td>
<td>22.5%</td>
</tr>
<tr>
<td>Wrist</td>
<td>40.0%</td>
<td>10.0%</td>
</tr>
<tr>
<td>Index finger</td>
<td>10.0%</td>
<td>26.9%</td>
</tr>
<tr>
<td>Ear</td>
<td>27.5%</td>
<td>18.5%</td>
</tr>
</tbody>
</table>

Table 7.1: The success rate of the measurements
The goal of this project was to research the feasibility of measuring the heart rate using a temperature based system. Although the available information on this topic is very concise, it can be concluded that it is possible to obtain the heart rate using this method with the use of an easy and cheap system.

In Chapter 6 a design was introduced, which led to reasonable results, described in Chapter 7. From these results it is possible to extract heart rate (requirement 3.1). In Chapter 7 it was also concluded that the system can measure the whole range of heartbeat signals (requirement 3.2). The total power consumption is 4.48mW. This is ~10%, and thus requirement 3.3 is also met. The remaining two requirements relate more to the possible further development and integration of the system. The system is definitely affordable for consumer electronics, and the sensor is already small enough to be integrated into a wired headset, as can be seen in Figure C.1.

Summarising these results implies that it is possible to implement a low power passive sensing module that uses a single measurement location. This is not possible with most established sensing methods such as PPG and ECG. However the acquired heartbeat signal contains less information than these methods.
The system can measure human heart rate. However, it is not able to deliver the desired result consistently. This chapter will contain improvements that can be made to the system, as well as recommendations for future research.

Our operating region is on the lower end of the spectrum. As mentioned in section 5.3, the noise in this region is higher due to flicker noise. This problem can be solved by shifting to another working region. This could be implemented by supply the bridge with an AC signal instead of DC. However, this modulation does require demodulation after amplification. In the case of the INA126, the noise will be reduced by one order of magnitude [17].

In section 6.1, the idea of using 4 NTC thermistors in the bridge was mentioned. If this is implemented correctly, more amplification would be possible, and a higher SNR could be achieved. However, this would require further thermal calculations and precise manufacturing, in order to achieve the perfect balance between thermal isolation and thermal conduction. Another option would be to research the effect of using a larger package NTC with the same $\beta$-value and $R_0$ instead of thermally isolated NTC’s. This could simplify the manufacturing process, but possibly add to the size module and costs.

It was also realised that the environment of the setup could have been better (e.g. by eliminating air-conditioning and other thermal noise sources). This would allow for the possibility of only changing one parameter at a time. This would allow an accurate analysis of motion on the performance of the system, which was not researched in this project.
• BPM - Beats Per Minute
• ECG - Electrocardiography
• FFT - Fast Fourier Transform
• GPS - Global Positioning System
• IR - Infrared
• NTC - Negative Temperature Coefficient
• PCB - Printed Circuit Board
• PPG - Photoplethysmography
• SNR - Signal to Noise Ratio
• USB - Universal Serial Bus


In order to test whether the circuit meets the required resolution of 0.2mK, a test set-up was used to validate if this requirement is met.

A.1. **THEORY**
In order to validate the resolution of the temperature of the circuit, a TEC1-12706 thermoelectric Peltier cooler was used[21]. This device transfers heat from the one side of the device to the other, controlled by a current. The direction of the temperature transfer depends on the direction of the current. The Peltier is controlled by a function generator connected to a MOSFET and a power supply with adjustable current output. In order to create small temperature fluctuations the current is lowered and the frequency increased until the output no longer changes at the given frequency.

A.2. **SET-UP**
By placing the sensor at the exact same position while testing different designs, it is possible to view quantitative differences between them. The sensor can be placed directly against or hanging over the Peltier cooler to simulate being placed directly against a human vein or in the proximity of a vein respectively.

*Figure A.1: Temperature probe above Peltier cooler for indirect temperature measurement.*
Figure B.1: The current BackBeat FIT
C

SENSORS

Figure C.1: Sensor integrated into a wired headset.

Figure C.2: Sensor used during measurements

Figure C.3: Amplification circuit