

An investigation into stress levels of final year undergraduate students using photoelectric plethysmography to detect heart rate variability

by Matthew Hunt

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**An investigation into stress levels of final year
undergraduate students using photoelectric
plethysmography to detect heart rate variability**

by

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Project submitted as part of the requirements for the award
of BEng (Hons) Computer Systems and Electronics

April 2016

DECLARATION

I, Matthew James Thomas Hunt declare that I am the sole author of this Project; that all references cited have been consulted; that I have conducted all work of which this is a record, and that the finished work lies within the prescribed word limits.

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Abstract

Photoelectric plethysmography has been used to develop small, wearable pulse rate monitors often using infrared or near infrared light. Such a device offers a simple, reliable and low-cost means of monitoring a person's pulse rate in a non-invasive manner. This report contains a comprehensive research and review of related literature, such as photoelectric plethysmography and electrocardiography concepts, infrared optoelectronics, Beer-Lambert's law, and stress and heart rate variability. The prototype proposed in this project is the development of a finger-tip PPG reflective device using infrared light, capable of obtaining data relating to arterial pulsation and heart rate variability. Average BPM and stress levels are calculated through an 8-bit microcontroller and displayed using an LCD after passing a system of signal conditioning to both amplify and filter the original PPG signal. The system is designed in an attempt to obtain empirical data to answer the hypotheses proposed.

TIA Index – Photoelectric Plethysmography (PPG), Electrocardiography (ECG/EKG), Heart Rate Variability (HRV), Beats Per Minute (BPM), Liquid Crystal Display (LCD), Light-Emitting Diode (LED), Pulse Width Modulation (PWM), Galvanic Skin Response (GSR).

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Chapter 1

Introduction

Stress related illness is a common occurrence in our modern times. Work-related stress accounts for 35% of all work-related ill health causes and 43% of all working days lost during 2014/15 in the UK [1]¹. Although short-term stress can be beneficial, the effect of long-term stress can have serious health repercussions both physically and psychologically.

With what seems an infinite number of preventative measures from regular exercise to social engagement, with each producing subjective results, it would seem straightforward to combat stress by using such techniques. However, difficulty ensues when attempting to determine how much stress is being placed on the body if any.

Stress is a notoriously difficult term to define where until very recently it was simply misunderstood. In fact, the term “experiencing physiological, emotional, or psychological stress” [2] did not appear in the Oxford English Dictionary until as recently as 1993. Thankfully due to our rapid expansion in science and engineering, extensive research has been done into the physiological effects of stress, and our understanding has drastically improved.

By using electrocardiography and heart rate monitors, research has shown that stress levels can be determined through measurements of RR intervals within the time domain of a beating heart. Discrepancies in the time domain known as heart rate variability are shown to have a direct correlation to stress. Where electrocardiography has been the defining method of obtaining electrical activity of the heart, a growing trend of using optoelectronics and photoelectric plethysmography has become popular during the past decade due to its non-invasive nature of measurement.

The purpose of this project is to produce a comprehensive paper of background research into the validity of photoelectric plethysmography in comparison to electrocardiography. It will determine whether information regarding the heart's electrical activity can be obtained using this trend. Thus, develop a prototype heart rate monitor using photoelectric plethysmography to determine heart rate variability and the levels of stress of final year undergraduate students.

¹ See **Appendix 1.0** for statistical analysis of work-related stress.

1.1 Aims and Objectives

The overall aim of the project was split into answering two hypotheses;

- *Can photoelectric plethysmography be used to determine stress through heart rate variability?*

And, if so;

- *Are final year undergraduate students under increasing stress levels to avoid the risk of under-achieving?*

Understandably, answering both hypotheses is a momentous task and may produce inconclusive conclusions with the limited time available. However, the primary focus to begin with will be answering the first hypothesis proposed; to decide if PPG measurements can be taken to determine heart rate variability and its relation to stress. To achieve this a series of goals and objectives must be met for the project to be conclusive in its final argument and to survive any criticism of the work.

The major objectives of the project are as follows;

- Produce an extensive and comprehensive literature review of related research, particularly in the field of heart rate monitoring, electrocardiography and photoelectric plethysmography and their relation to stress.
- Conduct primary research with questionnaires to obtain a consensus on stress levels of final year undergraduate students, and the marketability of a stress monitoring device.
- The design of a prototype portable device capable of obtaining information regarding the hearts cardiac cycles through the use of photoelectric plethysmography.
- Hardware and software implementation of said designs with sufficient and rigorous testing ensuring functionality of the proposed system.
- Obtain empirical data from final year undergraduate students at the University of Wales Trinity Saint David in regards to their levels of stress.
- Conduct thorough statistical analysis of the obtained data in order to provide primary research to aid in answering the second hypothesis.

Chapter 2

Research and Review of Literature

This chapter includes a comprehensive element of review and scholarly research, giving you as the reader sufficient background knowledge of main topic areas to assist in the understanding of any design decisions made for the system. Relevant subjects covered include heart rate monitors; specifically electrocardiograms and plethysmographs, photoelectric plethysmography concepts and fundamentals, heart rate variability and its relevance to the measurement of stress, and how light interactions with the skin can differ down to skin anatomy characteristics and Beer-Lambert's law of absorption.

2.1 Introduction to Heart Rate Monitoring

In recent history, particularly during the 21st century, multiple heart rate monitoring techniques have been employed to produce an extensive amount of heart rate monitors that are not only common in the health sector, but more recently in sport and fitness industry.

Within the grand scheme of things, heart rate monitoring has only been around for a short time frame. On July 11th, 1951, a paper was published claiming *"There has long been a need for continuous information about cardiac action during surgery."*[3] In this paper, named The Cardiotachoscope by Himmelstein et al. a device which they named a Cardiotachoscope was described to record pulse rate instantaneously using film recording with the intent of use in surgical procedures. Although the device was successful by being able to display a visual image of pulse rate on cathode ray tubes of the electrocardiogram, the device contained an annoying artefact produced by the presence of instruments in the thoracic cavity. This artefact meant any procedure performed in the chest of a patient would result in sharp spikes which would mask low amplitude complexes resulting in interference with the operation of the device. Such artefacts are common among heart rate monitors, which are discussed later in this chapter.

A common misconception among heart rate monitoring are the different methods of heart rate monitors, where terms can sometimes be misused and misunderstood. Methods of heart rate

monitors often come with the answer of chest strap monitors, wearable wrist monitors or a device in which the subject places a finger on a particular spot to generate a reading. Although these are common types, it doesn't specify what method is used to produce a heart rate result.

The most common method of heart rate measurement is electrocardiography, where innovations in technology during the first half of the 20th century led to the 12-lead electrocardiogram. Plethysmography is the alternative option; which is the detection of blood volume changes within the microvascular level of human tissue. The sections following highlight the concepts and functionality of each method of measurement.

2.2 Electrocardiogram

Often abbreviated as an ECG or EKG, an electrocardiogram is a device capable of recording the electrical activity of the heart. Electrocardiography today is an essential part of the initial evaluation of patients, particularly when complaints of cardiac difficulties arise. The introduction of x-rays in 1895 and the electrocardiogram in 1902 provided much-desired information about the functionality and structure of the heart. In 1902, the first electrocardiogram consisted of a string galvanometer [4] (also known as Einthoven galvanometer), designed by Dutch physician William Einthoven. It was the world's first reliable electrocardiogram and the earliest instrument with the capability of measuring small electric currents that are produced by the human heart.

Modern electrocardiograms are now found extensively throughout medical practices and more recently within the sports and fitness sector, and are the most common type for obtaining electrical activity of the heart to provide an assessment of cardiovascular function. In the conventional 12 lead ECG, ten electrodes are placed on the patient's limbs and the surface of the chest. The magnitude of electrical signals produced by the heart is then determined by the twelve different leads (or angles) and are recorded over a set period.

The human heart is a complex biological engine, and like any engine, it requires a substantial source of fuel; such as oxygen. It is segmented into four major chambers with a large quantity of interconnections or nodes. These chambers include the left atrium, left ventricle, right atrium and the right ventricle. The contraction of the atria and ventricles are responsible for generating an electrical pulse, which is then altered based on the various muscles tissues or nodes between the chambers.

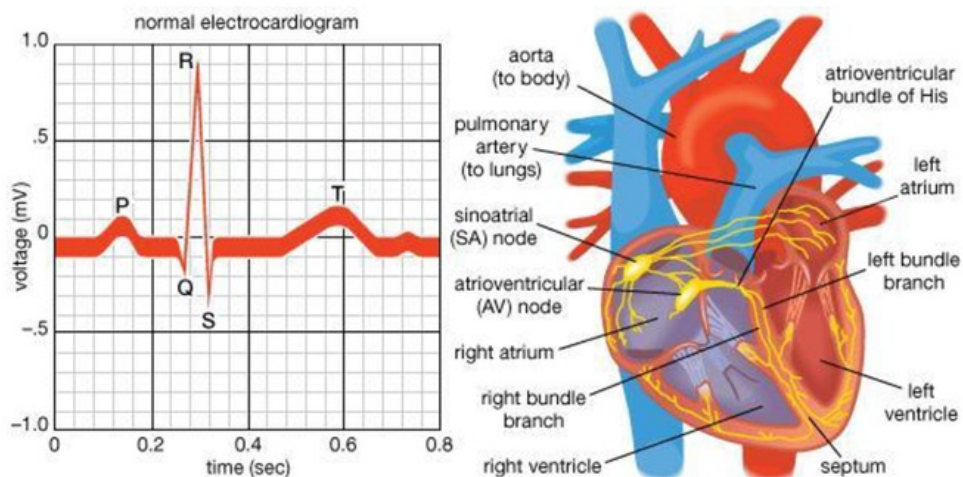


Figure 1; electrocardiogram signal and the human heart [5].

The image above taken from Encyclopaedia Britannica shows a typical electrocardiograph signal, which is displayed on a simple x/y graph where the x-axis is equal to time (in seconds) and the y-axis is equal to the voltage (usually visualised in millivolts). It also shows an image of the human heart, outlining the major components and nodes between chambers.

The sinoatrial node (SA) of the heart is where the electrical signal is first generated, which then moves through the two atria (left to right) and into the atrioventricular (AV) node. The AV node delays the signal and then sends it through the bundle of His which cause the ventricles to contract. By placing electrodes on the surface of the skin at specific angles in relation to the heart, it is possible to read the electrical signal produced using the electrocardiogram. This signal is then visualised on an x/y graph as mentioned and seen above. Different segments of the signal are represented and adjust based upon their position within the heart. The three main segments include the P wave, the QRS segment (known as QRS complex), and the T wave.

2.2.1 P wave

Preceding the QRS Complex is the P wave; which represents atrial depolarization. The cardiac cells within the SA node depolarise and perform an action potential in the upper right atrium of the heart as seen on the image in section 2.2. As the signal spreads through the right and left atria via the atrioventricular (AV) node and bundle of His, it causes them to contract. This segment of the electrocardiograph wave represents said contraction.

2.2.2 QRS Complex

The most prominent variation within the graph and the most commonly associated with a heart rate monitor is that of the QRS Complex. Although commonly referred to as one name, the QRS Complex consists of three waves; the Q wave which is a downward deflection, the R wave an upward deflection, and the S wave another downward deflection [6]. Notice that the QRS Complex, particularly the R wave, is of much higher voltage than any other segment of an electrocardiogram signal. Its size is due to the QRS Complex representing the contraction of both the right and left ventricles of the heart; where the chamber is bigger and tissue much thicker than the atria causing a large voltage.

2.2.3 T Wave

The T wave is when the cells of the ventricle repolarize. The ventricles empty during this stage and the valves between them and the atria close.

2.3 Plethysmograph

Derived from the Greek root 'plethysmo' meaning to increase and 'graph' which is the word for write, Plethysmograph is an instrument mostly used to determine variations and changes of volume within the human body (such as air or blood circulation). It is often used as an assessment of cardiovascular function by means of the peripheral arterial pulse, particularly with the increase of sensor technology and the expansive reach of microcontrollers in the last two decades.

There are four primary methods of plethysmography that are common today, each type capable of measuring volume changes in a unique way. The methods include air-displacement, impedance, strain-gauge and photoelectric with each method dependant on the type of transducer used.

Air-displacement plethysmography is the method of measuring the volume of an object by the use of measuring the volume of air displaced within a closed chamber. For this to work to measure heart rate characteristics, the subject enters an enclosed chamber which displaces a volume of air that is equal to the volume of the subjects' body mass. The volume is then dictated by

subtracting the remaining volume of air inside the chamber whilst the subject remains in the chamber from the amount of air in the chamber after the subject has left said chamber.

An impedance based plethysmograph applies a weak current through the limb of a subject, where the electrical resistance to the current flow is measured. By having multiple electrodes connected to various points of the limb and having the current flow from one point to the next to determine resistance. Such a device is proposed by Lawler, C. et al. [7], where the device uses cellulose tape stripping to separate the electrical behaviour of the surface of the skin. This method is the basis for galvanic skin response, where the conductance of skin is measured using the same technique and can be related to stress where the skin contains more moisture under a stressor, resulting in an increase in conductance.

Strain gauge plethysmography works by stretching the strain gauge and causing a decrease in diameter and therefore in voltage. It uses a transducer that is filled with mercury or an indium-gallium metal alloy conductor. The gauge provides a circumferential measurement by wrapping the device around the limb of a subject; when the limb in question either contracts or expands, it adjusts the volume of the strain gauge.

Finally, photoelectric plethysmography which concerns itself with the blood flow and blood volume of the skin. By using a transducer consisting of an LED (within the infrared spectrum of light) and a phototransistor, whereby the LED transmits light to an area of the subjects' limb (such as the finger). The light is then picked up via the phototransistor. Variations in blood volume result in variations of light being absorbed. Photoelectric is the method of plethysmography that this project holds the basis of its foundations.

2.5 Photoelectric Plethysmography (PPG)

Photoelectric plethysmography, also known as photoplethysmography and as its acronym PPG (which is the abbreviation used in this paper), is a non-invasive method to detect cardiovascular pulse waves that propagates through the body using a light source and a detector [8] (often referred to as a pulse oximeter). PPG is convenient, simple and an economically efficient method of gathering information on a person's cardiovascular system. The key thing to note with PPG is that it is non-invasive, meaning it does not need direct contact with the skin surface like that of other plethysmograph and ECG methods.

A technique of optical measurement that uses an invisible infrared light sent into the tissue and the amount of backscattered light represents the variation in the blood volume. This relationship was first discovered in 1938 by Alrick B. Hertzman in [9], where he proposed an argument showing the relation between the amount of absorption of light varies with blood content of a limb, in this case, of the fingers. However, with limited technology and advances in science and engineering Hertzman's research contained sources of error; particularly on the accuracy of the plethysmograph. There is a growing trend in the use of this technology as seen in [10], [11], [12], where information related to the cardiopulmonary system (heart rate and respiratory rate) is obtained.

Most changes in blood flow occur within the arteries and arterioles of human tissue [13]. For instance, during the systolic phase of the cardiac cycle (where the atrium contracts, allowing blood to flow from the atrium to the ventricle) more blood flows through the arteries than during the diastolic phase (where the heart is relaxed and not contracting). PPG sensors usually consist of an IR-LED (infrared light emitting diode) and a photodetector and detect changes in the blood volume of the artery either via transmission through or reflection from the subject's tissue. The IR-LED illuminates a section of the skin, where the emitted light is scattered in the tissue and partly absorbed by the haemoglobin in the blood. The light that isn't absorbed then emerges through the skin, where it is detected by the photodetector. The photodetector can be placed in one of two positions; either beside (reflection) or opposite (transmission) the light source. The light detected by either reflection or transmission is then converted into a voltage and its intensity dependent on a number of factors. Skin, connective tissue such as muscle, bone and of course the amount of blood in the vascular system are all factors of consideration; where blood is the property with a higher absorption coefficient. The variations in blood flow allow for variations in the amount of backscattered light to the photodetector. These changes allow for information to be gathered about the cardiovascular system of the subject.

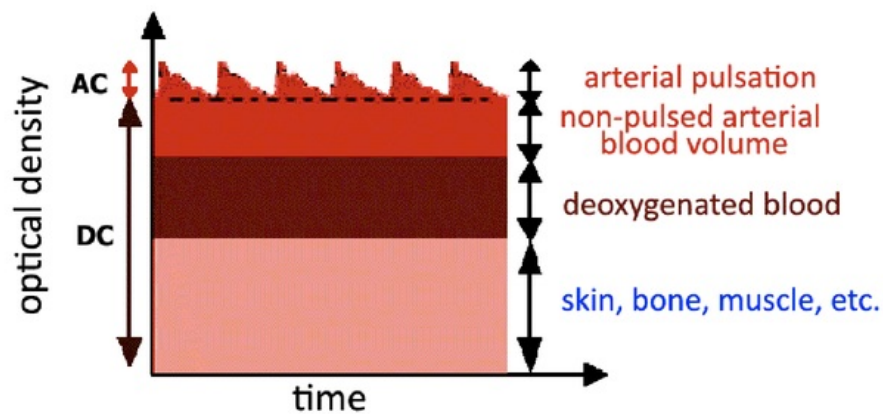


Figure 2; PPG signal formation as seen in [14].

The image above shows an example of a plethysmographic waveform; consisting of alternating current (AC) and direct current (DC) components. The AC component of the PPG waveform sits on top of the DC component as displayed above. The DC component of the waveform represents transmission or reflection of the optical signal from the tissue (such as skin, bone, muscle, etc.), and the average blood volume of deoxygenated blood (also known as venous blood) and arterial blood volume. The AC component sitting on top displays changes in blood volume during the two phases of the cardiac cycle; the systolic and diastolic phases.

As noted previously, PPG has two methods of measurement; transmission and reflectance as shown in the image below. Using the transmission method, the photodetector is placed on the opposite side of the LED source as seen on the left where it detects the remaining light after passing through the tissue. Whereas reflectance places the photodetector adjacent to the LED, which detects light that is reflected (backscattered) from the subject's tissue.



Figure 3; Transmission and reflectance component positioning [13].

For the transmission method to be an effective source of measurement, the placement of the sensor must be in an area where transmitted light can be readily detected; common locations include fingertip, earlobe, and nasal septum. The reflectance method can be viewed as a more practical

option; where component placement isn't of such importance (as the LED and phototransistor need not be opposite one another). Both methods are capable of detecting variations in arterial pulsation but often come with a questionable accuracy due to any additive artefact of the signal, such as undesired high frequency, low amplification and motion.

Area of measurement and location of the pulse oximeter is an important factor; affecting the quality of the signal and the protection against motion artefact. Although pulse oximeters are commonly found using fingertip measurements, recent research has delved into other locations and their comparison to fingertip PPG measurements. Proposed locations include the wrist [15], where Lee et al. produced a device capable of obtaining efficient PPG signals by measuring the arterial pulsation within the tibial artery, medial plantar artery and from capillaries in the fingers. Another shown in [16] looked at deriving cardio-vascular data from PPG measurements taken on the brachial artery. One clever innovation proposed by Shin et al. [17] combined the use of a generic headset capable of playing music that contained a transmissive PPG sensor placed within the ear cavity, so that arterial pulsation could be detected during physical activity.

The quality of the PPG signal is dependent on a number of factors; obtaining an accurate reading and deciphering the information being the biggest challenge. We mentioned previously that such factors could result in several types of additive artefact. This may affect the extraction of desired features and therefore the overall diagnosis. One such challenge is that of motion.

When attempting to obtain a PPG signal from a subject, it can be a difficult challenge to have said subject remain perfectly still. As a pulse oximeter obtains its data from the motion of blood moving within a vessel, it seems obvious that any motion of the vessel can result in incorrect data. As discussed and proven by Elgendi in [11], the cause of motion artefact is often assumed to be due to vibrations or movement of the subject. To help combat this issue, having the subject remain still and relaxed while maintaining a stable posture is essential when taking any measurements. This can be a difficult challenge in the proposed methods mentioned previously, where the devices particularly look at being able to obtain PPG data without restricting the use of fingers such as in PPG fingertip measurements.

Another obstacle found when taking PPG measurements is amplitude. Generally, without amplification deciphering information from a PPG signal is extremely difficult albeit impossible. Before any amplification, the PPG waveform is often only millivolts in amplitude potentially as little as 10mV. Signals as low as 10mV are extremely difficult if not impossible by most microchips to decipher. Therefore, significant amplification of the original signal would be necessary to overcome the issue for any hope of obtaining information from any arterial pulsation.

2.6 Skin Anatomy

The skin is the largest organ of the human body, comprising of about 15% of the total body weight of an adult. Performing many vital functions, it protects against external physical, chemical and biological threats, as well as preventing water loss from the body and acts as an insulator. The integumentary system is an organ system consisting of the skin, hair, nails and exocrine glands. Although the skin is only a few millimetres thick, it is by far the largest organ in the human body.

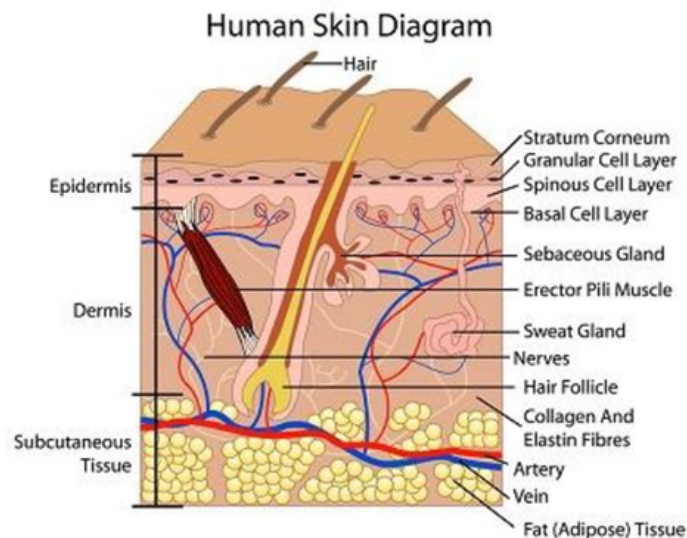


Figure 4; Human Skin Diagram [18].

Human skin is primarily composed of three primary layers, each with several other layers which have specific functions. The primary segments include the epidermis, dermis and subcutaneous tissue.

The epidermis is the outermost layer of the skin and is composed of five sub-layers [19]. The outermost layer of the epidermis is the Stratum Corneum, made up of keratin proteins and lipids. Secondly is the stratum spinosum, held together by prickly cells where keratin (protein) is produced. The third layer, named the granular layer, is where the precursors to keratin can be found that are responsible for the colour of skin and any pigmentation that may occur. The stratum lucidum layer is fragile, and is only found in specific areas of the body; usually the soles of feet and palms of the hand. The last layer consists of an area where keratin is produced in order to replace dead cells on the uppermost layers; this is the basal layer.

Dermis is the area between the epidermis and subcutaneous tissue and is composed of two layers; the papillary and the reticular layers. The reticular layer contains thick connective tissue, and is a layer that provides the skins durability and toughness. The papillary is composed of collagen and elastin, also connective tissues, responsible for keeping the skin flexible.

The final segment is a layer of fatty tissue known as the subcutaneous layer (or often as hypodermis). As with most fatty tissue, its primary purpose is to insulate [20] and to aid in metabolism. Cellulitis, a common and harmless deformation of skin is caused by the inflammation of this tissue [21].

2.6.1 Beer-Lambert's Law

The Beer-Lambert law also known by a number of terms relates to the attenuation of light and the properties of the material through which it is travelling. It is then in a sense a law of absorption, commonly found within chemical analysis, physical optics and mathematical physics. The law was proposed by 18th-century French mathematician Pierre Bouguer, who in 1729 published his works of "*Essai d'optique sur la gradation de la lumière*". Bouguer proposed a formula that was to define the quantity of light lost by passing through a given extent of the atmosphere, which is now commonly known as the Beer-Lambert law.

When considering a beam of light on a material, one of three things can take place; it can be scattered, absorbed, or transmitted. Transmitted light emerges from the material in the same direction as the light entering the material. Absorbed light is energy that is absorbed in the volume of the material. Scattered light is when a beam of light emerges in a different direction from its source.

This absorbance can be seen and measured using Beer-Lamberts law. In optics; the portion of physics dealing with the properties of light, light is measured in intensity (I) based on a number photons at a given point in time. When light passes through a solution of substance of a particular material, the intensity of light is measured before (I) and after (I°) traversing the material. Where I° is less than the value of I, energy is lost during the passing of said material [22] proving a level of absorption. This absorbance of energy is usually abbreviated as A.

The formula proposed using Beer-Lamberts Law to determine the level of absorbance is given by;

$$A = \log_{10} \frac{I^{\circ}}{I} \text{ [22].}$$

2.6.2 Light and Skin Interaction

Complexity ensues when looking at the relationship between light and its interaction with any biological tissue. However, a paper present in 1981 [23] by Anderson et al. investigated the penetration depth of light within human skin. It concluded that red and near-infrared light pass easily, resulting in IR wavelengths commonly being used as a light source for PPG sensors. In the visible region of the electromagnetic spectrum, the shorter wavelengths of light are found to be strongly absorbed by melanin, with water absorbing light in the ultraviolet region.

Referring to the diagram seen in section 2.5, one can see that blood (both oxygenated and deoxygenated) absorbs the largest quantity of light in comparison to other tissues such as bone, muscle and skin. Therefore, variations in the amount of blood present within the biological tissue are directly proportional to the intensity of light detected by the photodetector. Penetration depth, however, is determined based on both wavelength and distance between the photodetector and the source of light (LED). The green-yellow spectrum of visible light (some 500-600nm in wavelength) proves to have the largest modulation depth of pulsatile blood absorption, and green light more accurate for measuring blood flow in the skin (superficial flow, where blood does not travel directly through an artery). Infrared wavelengths (or near-infrared) are the most popular form of light source for optical plethysmographs due to IR wavelengths being ideal to measure deep tissue blood flow, such as flow through muscular tissue where the arteries lie.

In recent years, investigations into using the light of other wavelengths have become more common; particularly that of green light. As seen in [24], [25] and as a commercially viable product using an EKG in Mio Global' Alpha 2 [26], green-light based PPGs are becoming a very real possibility. However, using infrared or near-infrared wavelengths to detect arterial pulsation has extensive research backing its theory; it is a proven method of measuring the cardiopulmonary system and to obtain data regarding heart rate characteristics.

2.9 Electromagnetic Spectrum

What is commonly referred to as light, is very likely visible light; which is the electromagnetic radiation related to a specific set of wavelengths within the electromagnetic spectrum. The term light is used to describe the section of the electromagnetic spectrum ranging from radio waves to gamma rays. *"Electromagnetic radiation waves, as their names suggest are fluctuations of the electric and magnetic fields, which can transport energy from one location to another."* [27]. Light is

then, as the definition above states, a form of energy. This energy travels in the formation of a stream of photons, which is the smallest quantity of energy. The distance of a ray of light (measured from peak-to-peak) is usually measured in nanometres (nm).

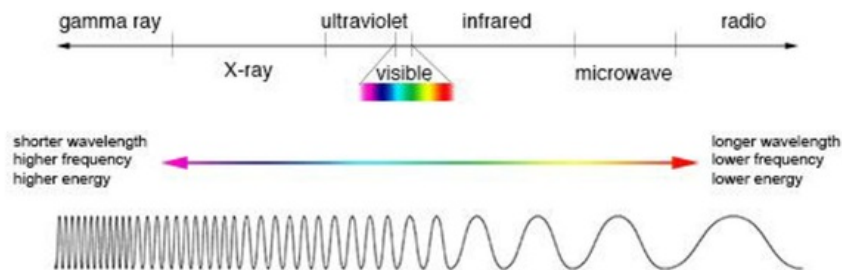


Figure 5; electromagnetic spectrum [28].

2.7 Infrared optoelectronics

Optical science, engineering and technology have grown rapidly in the last decade with optical engineering emerging as an important discipline in its own right. Modern day optoelectronics, defined by J. Wilson et al. as “*the interaction of light (in the wavelength range from about 100nm to 2um) with matter in gaseous, liquid or solid form, and the devices which depend on these interactions*” [29]. This definition is, of course, broader than that, where restrictions are often placed on the topic based on what device(s) the light source interacts with.

In this case, we look at the use of infrared optoelectronics. Infrared light is a light source whose frequency is just below the range of visible red light, specifically a frequency which falls between 1 to 400 THz. The infrared spectrum falls right between microwaves and visible light. Just like visible light, infrared light can be blocked by solid objects and is capable of bouncing off of reflective objects or surfaces. In modern engineering and technology, it has become commonplace that infrared light be used to detect objects that we can't see in visible light, infrared can be manipulated to form a number of applications such as night vision or wireless communications.

An infrared source typically takes the form of an infrared light-emitting diode or IRED. It is known by a number of acronyms, IR emitter, LED, IRED and so forth but IR-LED will be the term used throughout this paper.

2.8 Introduction to Stress

A commonplace term, stress is often used on a daily basis in reference to work-overload, relationship issues amongst many other day to day occurrences. Stress plays a key role in our daily lives, often leading to a decrease in our performance of everyday tasks as an individual such as work, sleep and eating. It also affects our health, capable of leading to disorders such as anxiety, depression, obesity and anorexia. Although this seems like common knowledge, what is stress? What happens to our bodies for us to develop such damaging and detrimental health issues?

Stress can be defined as the process by which we perceive and respond to certain events, called stressors, which we appraise as threatening or challenging. The key word here is perceive; our perception, or the way in which we interpret information and the meaning that we give to it, is key to what triggers stress. This perception is dependent on an individual basis, where one person may perceive an event as a threat, another person may see it as a challenge and rise to the occasion. Although in both cases said event is defined as a stressor, it can have both beneficial and negative effects on a psychological and physiological level.

Particularly when short-lived, stress that we perceive as a challenge can be beneficial and has the potential to provide a number of benefits; improve creativity, temporary boost to the immune system, develop resilience to stress and even provide motivation. However, long-term stress is mostly associated with the negative effects that it brings. While there are numerous methods to combat stress, it is difficult to know how effective such methods are without an accurate measurement, where a subject may be under small amounts of long-term stress without even realising.

Heart rate variability (or HRV), a term used to describe variations in time intervals between heart beat pulses, has been associated with the functioning of the autonomic nervous system and its activity. The recent research proposed this century has investigated into these associations and whether heart rate variability can be used to determine various physiological conditions, in particular where the body is under stress.

2.8.1 Heart Rate Variability

With 25 years of clinical research results have shown that heart rate variability (HRV) is one of the most reliable indicators of determining stress. HRV relates to the variation in the time intervals between one heartbeat and the next. When considering heart rate the general assumption and first consensus is that of beats per minute (BPM), where it would seem that should BPM increase the time intervals between heartbeats would decrease. However, this is not entirely the case.

One would assume that a consistent and regular heartbeat would relate to a healthy heart, interestingly this statement is false. *"A healthy heart is not a metronome"* [30] produced in the Frontiers in Psychology journal by Shaffer et al. presented their findings on the dynamic balance between sympathetic and parasympathetic nervous systems and their responses. They showed that a healthy nervous system is consistent with generating the heart rhythm and modulating the time interval between heartbeats. In fact, this research determined that regular pulses with a lack of modulation in the time domain were associated with stress-induced states, health risks and higher mortality rate. Therefore it safe to assume that a lack of variations within the time domain of heartbeats can be measured to determine the stress levels of a subject.

A paper published in 2000 by Anishchenk et al. [31] from Saratov State University in Russia proposed a comparative analysis method to determine various states of the cardiovascular system while under stress using an electrocardiogram to measure RR-intervals of the heart. The research suggested was successful in its endeavour to a degree, showing that stress could, in fact, be measured through the time domain of RR-intervals. However, it was discovered that *"Nonstationary measurements may drastically complicate the interpretation of results obtained"* [31]. To combat this issue and to produce accurate measurements of stress using information from the time domain, measures would need to be taken to remove potential motion artefact from the signal; which could be achieved using various methods such as a consistent pressure of contact between subject and sensor or stability of the subject.

Chapter 3

Research Methodology and Market Research

According to the Cambridge English Dictionary, research is “*a detailed study of a subject, especially in order to discover (new) information or reach a (new) understanding*:[32]. Research can come in a number of forms, in particular, primary and secondary research. The individual performing conducts primary research, such as from surveys, interviews, and laboratory tests. Secondary research refers to information from studies previously performed that are used to obtain relevant information or to assist in proving a particular theory, concept or argument.

This chapter looks at the procedures followed to generate a solution to the proposed hypotheses, highlighting the research methodologies used throughout the project. It also provides a review and analysis of popular products that are commonly found on today’s market.

3.1 Research Methods

There are a number of strategies employed to undergo research to achieve promising information or to reach an understanding of a particular topic. One such method introduced by Saunders et al. is the research onion. The research onion is a model that “*provides an effective progression through which a research methodology can be designed*” [33]. The model does this by consisting of multiple layers that represent key aspects that require attention when conducting any research. The research onion model consists of 6 layers, each layer containing subsequent techniques and methods of research. The layers are as follows;

Layer 1; Philosophical Stances

Layer 2; Approaches

Layer 3; Strategies

Layer 4; Choices

the research question.”[34] It follows a logical process of reason where theory and research attempt to provide an answer to the original statement, backed by any data collected.

Layer 3 – Case Study and Experiment

The strategic layer of the onion model offers a wide range of potential techniques. Two of which are used as part of the project’s research methodology; case study and experiment. A case study is the process of record of research into the development of a particular group, person or situation. Collected data within a case study often includes watching aspects of the subjects’ behaviour and interviews with participants. The device proposed in this project does exactly that; whereby data is collected via a sample of final year undergraduate students to gather information on their levels of stress.

Experimental designs, the second method used as part of the strategic layer, are often more rigid and scientific in their structures that allow research to be replicated. Such strategies generate data that can be statistically analysed, greatly assisting in developing a conclusion to the proposed hypothesis.

Layer 4 - Mixed Methods

The fourth layer of the onion model refers to the choice between how the researcher wants to use quantitative and qualitative methods. Quantitative research is the data of numbers and values, concerned with quantity and measurements. Qualitative research is a far more subjective type; it is concerned with rich data such as opinions, descriptions, and personal accounts.

Neither quantitative nor qualitative research stands superior to the other throughout this project, as both methods are used equally. Quantitative research would refer to any data obtained from the proposed system such as average heart rate and heart rate variability. Where qualitative research would be represented by the subjective opinions, such as surveys undertaken or feedback of the subjects using the device.

Layer 5 – Time Horizon

Layer five determines the length of the project and design, the framework within which the project is intended to come to a conclusion. The research onion model looks at two methods of time horizons; longitudinal and cross-sectional. Cross-sectional designs use both methods of the choices layer, i.e., qualitative and quantitative research and take measurements at a single point in time.

Longitudinal designs are also capable of using both methods of research but have a longer period of study and measurement.

A longitudinal approach best suits the proposed project, where the fundamental work carried out can be expanded upon to further investigate the project's hypothesis and generate a more accurate conclusion.

Layer 6 – Data Collection & Analysis

This is where the researcher decides what data collection methods work best and the type of analysis used to produce the results for a conclusion to the proposed hypothesis or opening statement. Both primary and secondary data were collected and analysed throughout this project; where chapter 2s main focus lies on the analysis of secondary research, the later chapters of the design and implementation stages carry out primary research such as laboratory tests.

3.2 Tools and Applications

Throughout this project a number of hardware tools and software applications were used to perform particular functions. A full list of hardware and software applications used is listed below;

Hardware

- SJ Electronics Function Generator
- Tektronix TDS Oscilloscope
- PicoScope
- Breadboards
- Hand-held voltmeter/ammeter
- Soldering iron
- Picket 3

Software

- Proteus 8
- CCS C compiler
- MPLAB IDE
- Microsoft Excel
- Various Online Libraries (IEEE, NCBI)
- Survey Monkey
- Microsoft Visio

3.3 Market Research

Heart rate monitors come in various shapes and sizes, some present a wearable device such as a wrist band/watch or chest strap, others come in the form of an application for an android or IOS device, whereas some employ both methods. Although the majority of these devices (particularly wearable devices) come in the form of an electrocardiogram, more recent devices have been incorporating photoelectric plethysmography as their method of measurement. The major factor when comparing the differences between individual devices came down to cost vs application; where the cost starts to increase when more features are added to the device. Quality becomes another issue; where cheaper devices can often produce estimations and/or inaccurate readings, particularly those designed to monitor physical activity and fitness. See **Appendix 5.0** for current heart rate monitors investigated.

3.3.1 Questionnaire

In order to produce quantitative data through primary research a questionnaire was developed using the tools available by Survey Monkey. The questionnaire consisted of seven questions relating to stress and stress monitors in an attempt to gain an understanding of the general public's opinion on such device. The full questionnaire including graphs produced using the data obtained can be found in **Appendix 6.0** and are analysed in the final chapters of this report.

Chapter 4

Initial System Design

The following chapter discusses the processes taken during the design phase of the project, highlighting potential difficulties found and how to overcome them. This chapter looks at the initial design of the system and its requirements. However, the design phase in its entirety is spread across two chapters; where the next chapter focuses on the justification of component selection and the simulations of the system.

4.1 System Requirements

Before looking at the system in detail and investigating various components that hold potential, having an overview of the system and its requirements is essential. Decision making, particularly in the design phase, is of utmost importance with each decision requiring justification. Having a list of requirements and a model in which to use as a reference helps greatly in this regard. As highlighted in the introduction, the system will use PPG technology and infrared light to obtain information from heart rate; specifically average BPM and heart rate variability. To achieve this, the requirements are as follows;

PPG signal – obtaining heart rate data

- The first major segment of the system, the input. As can be seen when discussing the literature review, PPG signals are low in amplitude and susceptible to noise therefore require substantial signal conditioning.
- In order to amplify the signal, a series of cascading operational amplifiers will need to be used in a specific set-up to obtain the required gain. Without amplifying the original PPG signals, there would be no hope for the microcontroller to detect any input.
- With amplification also comes the unwanted ambience. In conjunction with the operational amplifiers, RC filters will be applied in specific configurations in order to minimize high frequency artefact on the signal.

The Microcontroller – making use of the data

- The heart of the entire system, a microcontroller (pic) will be used in order to detect the PPG signal and make use of its data.
- It will provide two main functions for the system; performing calculations in order to determine average heart rate (bpm) and measure heart rate variability, allowing for an indication of stress levels.
- This information will be displayed to a liquid crystal display in 4 bit mode. It will also control a series of LEDs as a visual aid to the system.

LCD and LEDs – a city of lights

- As mentioned, the microcontroller will display its calculations onto an LCD in 4 bit mode. A 2x16 LCD will be used, with the top row displaying average BPM in real-time and the bottom row an indication of stress, based on HRV calculations done within the microcontroller.
- A series of LEDs (6-8) will be used as a visual aid for the system. Each LED will be given a set threshold based on the input to the microcontroller, so that when a threshold is hit that specific LED will light up. The idea is to generate a smooth flow of LEDs in series upon detecting heart rate measurements.

An output – PWM HR data

- Although the three sections listed will be enough to answer the opening statement of the project, an additional feature may be included. Provided time constraints do not limit the systems potential, the microcontroller will be able to generate a pulse width modulation (PWM) of the stream of heart rate data being measured. This way the system could interact with a data-logging device or monitoring system, capable of storing heart rate measurements and / or displaying the data to a monitor in real time.



Figure 7; simple block diagram of system.

4.2 Block Diagrams and Flowcharts

Block diagrams are diagrams of a system which highlight principal segments and their functions which are represented by blocks. These blocks are then connected to each other using lines or arrows to show their relationships. They provide a simple and effective method of bringing clarity to a systems concepts, allowing for understanding before looking at detailed circuitry and schematics. These diagrams can also represent software logic which are known as software flowcharts.

This section lists a series of block diagrams and software flowcharts of the overall system, highlighting key segments and their relationships. Detailed explanations of each segment within the system are found in the coming sections of this chapter.

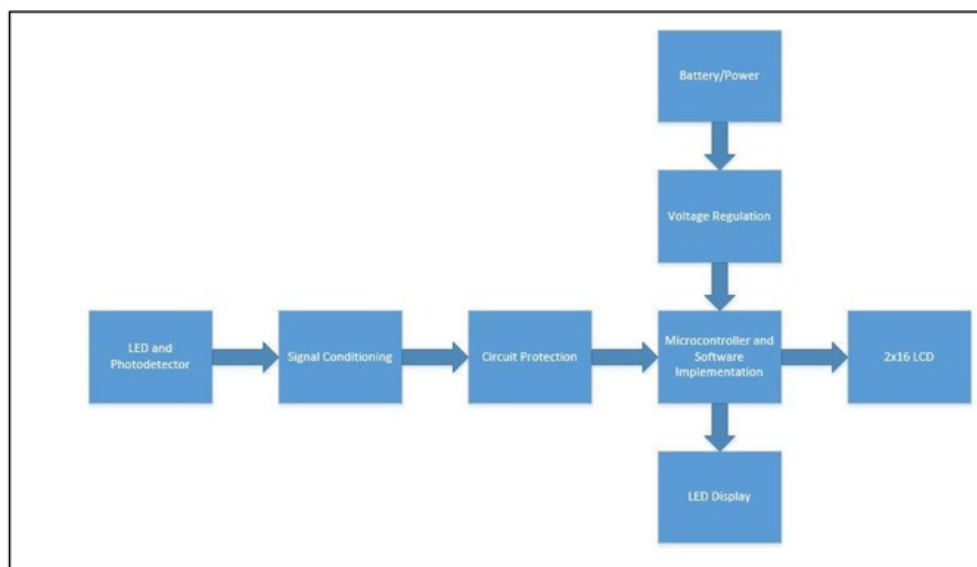


Figure 8; expanded block diagram of system.

The above image shows an enhanced block diagram expanding upon the basic diagram shown previously. This diagram highlights the key segments within the system and their relationships. Note that the LED and photodetector, 2x16 LCD and LEDs are also powered through the voltage regulator.

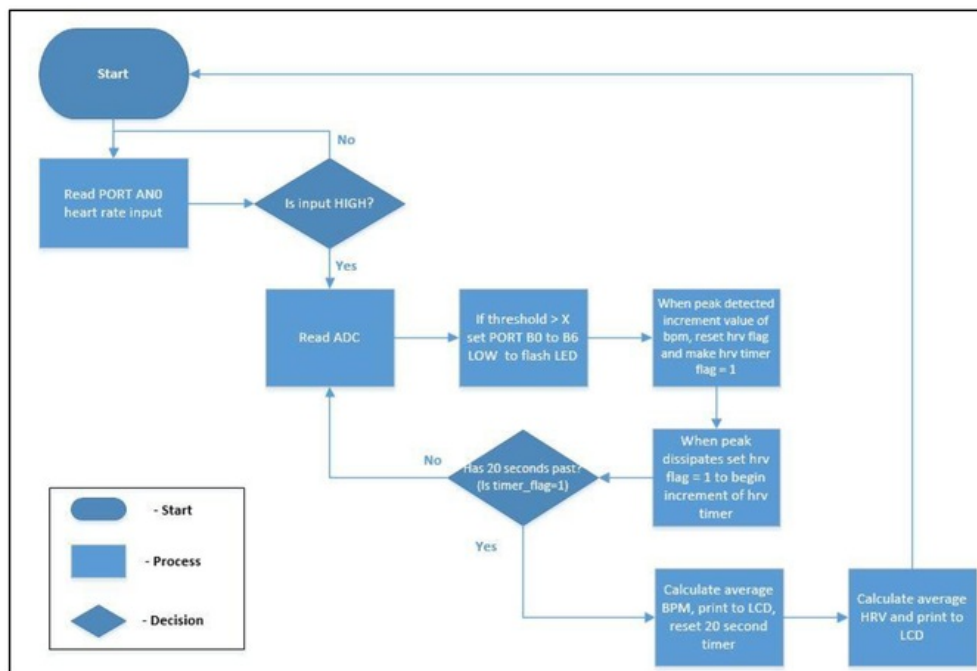


Figure 9; software flowchart of system.

The diagram above shows the conceptual logic of the systems software. Where the Start function is used to start the program, a Process is used to perform a specific action and the Decision performs an action based on a given parameter i.e. yes or no. The flowchart shown is a basic design of how the system would be implemented, starting by reading the analogue input on PORT AN0 and continue to do so until a signal is received.

4.3 Signal Conditioning

Before using data gathered from the PPG sensor to generate results, a series of signal conditioning techniques were considered. The conditioning of the signal included the use of RC filtering to remove high-frequency artefact while amplifying the signals significantly to be read by the microcontroller.

A signal conditioning system of multiple operational amplifiers in a cascading setup was considered with RC filtering in between, as to limit the increase of unwanted high-frequency noise. The system was separated into four stages; two stages of RC filters to ensure only amplification of the variations in AC arterial pulse by removing unwanted frequencies, and two stages of operational amplifiers to amplify the desired signal.

4.3.1 Amplification

Amplification, also known as amplification factor or gain, is the process of boosting or increasing the strength of a signal. A circuit designed to perform amplification is known as an amplifier. Amplifiers often consist of operational amplifiers and a combination of resistors that determine the level of gain. A well-known complication with PPG measurements is that of the size of the signal, usually only millivolts (mV) in size. The image below shows a pico-scope measurement of the sensor being used within the system before any amplification where the average voltage is 10 millivolts peak-to-peak (mV pk-pk), proving that significant amplification is required.

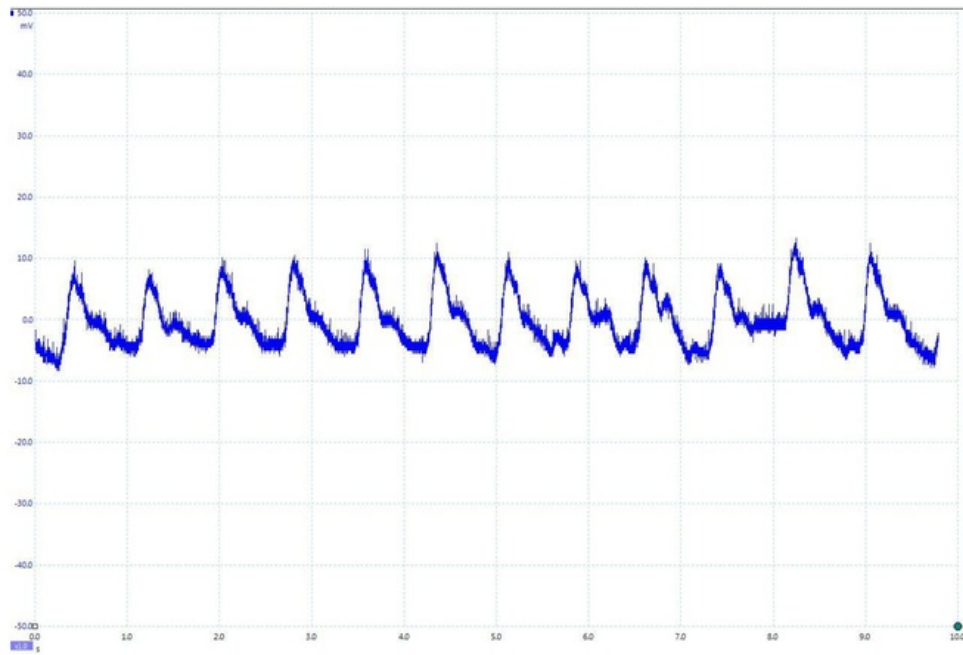


Figure 10; pre signal conditioning PPG signal.

The output of an amplifier can be determined in terms of an output/input ratio. The technical term for this ratio is known as gain. Gain is a unit-less measurement as it is a ratio of equal units' i.e. current in/current out or more commonly voltage in/voltage out. Gain is symbolised as the capital letter **A** (or **Av**) when referred to in a mathematical context. The formula to determine the value of **Av** is given by;

$$Av = \frac{V_{out}}{V_{in}}$$

With the PPG signal currently sitting at as little as **20mV** significant gain is required in order to boost the amplitude in the region of **1-5v**, allowing the microcontroller to detect a far more accurate reading. The ratio of **Vout/Vin** to **Av** only determines the amount of gain of that particular setup of an operational amplifier. The amount of gain produced by the operational amplifier is dependent on the resistor values present (also known as biasing). The formula used to determine how much gain is produced is given by;

$$Av = \frac{R1}{R2}$$

Where the resistor R1 is connected between the negative terminal of the amplifier and to ground, and R2 is the resistor between the negative terminal and the output terminal of the amplifier.

4.3.2.1 Gain Calculations

The following calculations were performed to determine the required amount of gain and resistor values for the PPG signal to reach an area suitable for the microcontroller to read accurately.

Gain calculations stage one – Using the formula;

$$Av = \frac{V_{out}}{V_{in}}$$

Where **Av** = increase in gain, **Vout** = the output of the amplifier, and **Vin** = the input of the amplifier. With the knowledge that the PPG signal amplitude average is **20mV** prior to any amplification and that we require a voltage input to our microcontroller of **1-5v**, using the above formula the gain required is determined by;

$$Av = \frac{1v}{20mV}$$

$$Av = 10$$

For the PPG signal to reach the minimum requirement of **1v**, the operational amplifier in the first stage requires a gain of at least **10**. If **Av = 10**, then the value of **R1** must be so that it is 10 times larger than that of the value of **R2**. The following resistor values were considered;

$$Av = \frac{10k}{1k} = 10$$

$$Av = \frac{20k}{2k} = 10$$

Note that achieving a gain of 10 has an infinite amount of resistor combinations, all of them reaching the desired gain. Although the gain remains the same in theory, there are downfalls with choosing a combination of resistors that are either too large or too small. Using small resistors requires the usage of a much higher current to provide the appropriate voltage drop for the operational amplifier to work. Most amplifiers are able to provide 10's of milliamps (mA) but the more current produced, the more heat that is generated, which could be problematic.

On the other hand using large resistors tend to cause issues with the non-ideal behaviour of an amplifier's input terminal. A universal assumption is made that an ideal operational amplifier has an infinite amount of input impedance, however, this is not the case in reality where there is always a finite current flow through the input terminals (no matter how small).

To combat these issues using resistor values of a middle range (in the 1000s of ohms) is often advised. Therefore, the resistor values of **20k** and **2k** were used in relation with an operational amplifier to achieve the desired gain of **10**. With a gain of **10** during the first stage achieved the PPG signal would average an amplitude of **1v**, *in theory*, however engineering practices in reality have a tendency to avoid following exact theory due to infinite possibilities.

For the PPG signal to have an average amplitude of **1-5v** another stage of amplification was required to ensure that should any discrepancies appear during the implementation resulting in the signal dropping below a **1v** average, the second stage would compensate for any loss of gain. The second stage of amplification would be a direct copy of the first, providing a gain of **10** using the same resistor values of **20k** and **1k**.

However, with the PPG signal obtaining an increased amplitude by a factor of 10 gain so will each of the undesired high frequencies on the signal. Therefore substantial testing of different variations of capacitor values to determine frequency cut off and capacitive reactance was required.

4.3.1 Filtering

Filtering is the means of removing undesired frequency or range of frequencies from a signal within a circuit. A circuit designed to perform frequency selection is called a filter. In order to perform this process of filtering, a number of steps need to be undertaken.

Step 1 – Select a frequency cut-off point. Frequency cut-off point refers to the -3db point where the frequencies after a certain point begin to attenuate, often labelled as **F_c**. The mathematical formula to determine said point is given by;

$$F_c = \frac{1}{2\pi RC}$$

Where **R** is the value of the resistance of the circuit, and **C** relates to its capacitance. In other words the resistor and capacitor values of the circuit. This formula represents the **F_c** of an RC (resistor-capacitor) filter, which are both easier and cheaper methods of filtering when using op amp configurations (in comparison to LC filters).

Step 2 – Determine value of resistor. This can be done by using a simple transposition of the formula to calculate **F_c** as seen below;

$$R = \frac{1}{2\pi F_c C}$$

If the desired frequency cut-off point is known, then using transposition of formula is an easier way of determining the required values of components. This can also be done to determine capacitor value, where R and C are swapped;

$$C = \frac{1}{2\pi F_c R}$$

Step 3 – Calculating capacitive reactance. When a voltage is applied to an AC circuit the polarity of the signal is in constant flux, alternating between negative and positive polarities determined by the frequency of the signal. The most typical example of this is found in a standard sign wave. These changes in polarity cause variations in the charge of a capacitor at a rate determined by the frequency. As the capacitor charges or discharges, its resistance changes which adjusts the flow of current passing through the capacitor; as of course current is directly proportional to the level of resistance. This internal resistance within the capacitor is known as capacitive reactance (measured in ohms) which is usually abbreviated as the value of **X_c**. The formula to determine the value of **X_c** is given by;

$$X_c = \frac{1}{2\pi f C}$$

The property of capacitive reactance results in capacitors being ideal components for use in AC filtering. This is due to the capacitor applying a signal path forcing unwanted frequencies to ground. For the proposed system to work it requires the use of RC filtering to remove high-frequency artefact, so an understanding of how the systems capacitive reactance is required. As DC signals are of zero frequency X_c can be seen as an infinite resistance (simply because there are no alterations in frequency), resulting in the rule that capacitors pass the AC component but block DC components of a given signal. This is an ideal characteristic for the proposed system; where the AC component of our PPG signal relates to changes in arterial pulsation.

4.3.1.1 Filter Calculations

Following the steps highlighted previously, the following calculations were proposed to determine frequency cut-off points and capacitive reactance of the two stages of RC filtering used within the system.

Frequency cut-off – Using the formula;

$$F_c = \frac{1}{2\pi RC}$$

Where **R** = resistance and **C** = capacitance. With the knowledge that the average frequency of the desired PPG signal will sit around at **1Hz**, the first stage required the -3db point to be above **1Hz** but low enough as to remove unwanted high frequencies. To determine a sufficient cut-off point, a number of resistor and capacitor values were considered. The frequency cut-off of the first stage of filtering was determined by transposing the above formula to calculate the value of R when using a 1uF capacitor;

$$R = \frac{1}{2 * \pi * f * 1\mu F}$$

Using the capacitive reactance formula of;

$$X_c = \frac{1}{2\pi fC}$$

It is possible to determine the reactance of a 1uF capacitor at varying frequencies.

At 1Hz = 159,154.94 ohms (160k resistor)

$$X_c = \frac{1}{2 * \pi * 1Hz * 1uF}$$

At 1.5Hz = 106103.29 ohms (10.6k resistor)

$$X_c = \frac{1}{2 * \pi * 1.5Hz * 1uF}$$

At 2Hz = 79577.47 ohms (80k resistor)

$$X_c = \frac{1}{2 * \pi * 2Hz * 1uF}$$

At 2.5Hz = 63661.97 ohms (64k resistor)

$$X_c = \frac{1}{2 * \pi * 2.5Hz * 1uF}$$

It is evident that an increase in frequency results in a decrease of internal resistance within the capacitor. This is exactly what is needed in the system. As current constantly follows the path of least resistance, by decreasing the internal resistance across the capacitor when frequencies are increased the path to ground becomes much more attractive for those frequencies. For example, if a frequency of 50Hz (typical frequency of mains power supply) spikes on the PPG signal then the resistance across the capacitor will be **3183.09 ohms**, given by;

$$X_c = \frac{1}{2 * \pi * 50Hz * 1uF}$$

3.2k resistance! That is a significant decrease in comparison to the **160k** resistance seen when **1Hz** signals are present, resulting in the **50Hz** artefact following the path to ground. Note that this theory applies to an RC filter in a series circuit. The RC filtering done with the use of an operational amplifier is somewhat more complex, where two resistors are in series with each other but the larger resistor forms a parallel circuit with the capacitor.

In an ideal circuit, placing the capacitor in parallel with the largest resistor would not have any significant effect on the gain of the operational amplifier. Sadly this is not the case when dealing with the physical world. Increasing the size of the capacitor would lower the frequency cut-off of the system, resulting in the attenuation of unwanted frequencies to a more accurate degree. However this would also result in a decreased gain which could cause significant errors and reduce the output voltage.

To determine the most suitable capacitor for the system using a 20k/2k resistor setup to achieve a gain of 10, the above theory was put to the test by calculating the amount of gain lost with an increase in capacitance. Creating a balance between frequency attenuation and loss of gain was crucial to producing an optimal PPG signal to be read by the microcontroller. This was achieved by applying the theory and creating excel spreadsheets to determine the ideal trade off².

4.4 Software Algorithms and Logic

With a suitable signal amplitude and the removal of unwanted high frequency ambience, the microcontroller would now come into play. The requirements of the microcontroller to achieve the objectives of the project included;

- Converting the analogue PPG signal into a digital representation
- Have a 20 second timer or interrupt
- Activate a series of LEDs when an input is detected
- Measure the amount of pulses and calculate an average heart rate (BPM)
- Print BPM to an LCD
- Measure time between each pulse and store in an array
- Calculate average time between beats over the 20 second period (HRV)
- Display HRV to an LCD

The basic design of how the system would flow can be seen on the software flowchart highlighted on **section 3.2**. Thankfully, using Embedded C and CCS' C compiler allows for a severe reduction in headache when setting up the desired characteristics of a microcontroller (in comparison to assembly language programming). By allowing the use of an interrupt it is possible to create a simple timer whilst using flags to incorporate it into various statements and expressions.

² See **Appendix 2.0** for tests performed on optimal capacitance.

Creating a visual aid for the user through using multiple LEDs could be possible by allowing for thresholds. If the ADC has an 8bit resolution (meaning a total of 256 samples of the analogue signal) then by using a threshold it is possible to develop a number of possibilities; such as a simple Boolean expression. Where if the threshold goes above a specific value (say 200, out of the 256 possible samples and 256 being the peak voltage of the analogue input) an LED could be set to turn on and vice versa.

Using this concept of thresholds can also be done to determine when a peak is being produced and when it is dissipating. For example; using two variables it is possible to determine whether the input voltage is increasing or decreasing, where the first variable associates with being above a threshold of 200 and the second variable associated with the amplitude of the signal prior to it reaching 200. Therefore it is possible to count pulses provided by the PPG signal when a threshold is reached, and also to increment a timer when the pulse dissipates until the next pulse.

By having the ability to know when pulses are being formed by the voltage increasing and decreasing, both heart rate measured in BPM (beats per minute) and heart rate variability can be determined. BPM relates to the amount of times a heart beats across a 60 second period, therefore by measuring the amount of pulses obtained from the PPG sensor over a 20 second period BPM can be determined by;

$$BPM = n * 3$$

Where **n** is the number of beats measured over 20 seconds. Heart rate variability on the other hand is a more complex procedure. The method proposed to determine HRV is to provide an average of variations between each pulse by incorporating the concepts of a threshold, ring counter, timer and flag. When a certain threshold is met (i.e. the pulse dissipates) a timer begins to increment until the next peak. The data measured from pulse 1 to pulse 2 is then stored in an array (or this case a ring counter) and stops counting whilst a pulse is present. The process then repeats each time a pulse dissipates i.e. drops below a certain threshold. An average could then be taken to determine heart rate variability by dividing the data stored in the ring counter by the amount of pulses measured i.e. if the time between 25 pulses were measured then average HRV can be determined by;

$$HRV = \frac{n}{25}$$

Where **n** is the total amount stored within the ring counter.

Chapter 5

Component Selection and System Simulation

The following chapter provides justification of components chosen and used within the system, comparing the various components considered. These justifications are split into five sub-sections for the main components of the system; the sensor, microcontroller, liquid crystal display (LCD), voltage regulation and choice of battery. Following comparisons of the components are simulations of the system done in Proteus, providing screenshots of key areas and highlighting any fundamental issues.

5.1 Sensor Selection

The first major component of the system is the method of obtaining data from changes of blood volume within the tissue of a subject. Reflective optical sensors were found to be the most appropriate approach. A sensor was selected for the device based on its characteristics; using an IR-LED emitting at a wavelength between 750nm and 1000nm and a photodetector capable of measuring the amount of backscattered light, detecting changes in the blood volume of the artery.

With the use of a reflection sensor where the backscattered light is proportional to changes of blood volume, an appropriate location of measurement was necessary. The most commonly used locations found in research were fingertip, earlobe, and the nasal septum. Fingertip measurements were chosen for the system based on its reliability (a strong arterial pulse) with an appropriate and convenient location whilst taking measurements. Three sensors were considered for use within the system and their characteristics are highlighted below.

5.1.1 NJL5501R

Introducing the NJL5501R reflective optical sensor, containing built in LEDs at the visible red light and infrared wavelengths and high sensitive photodetector.

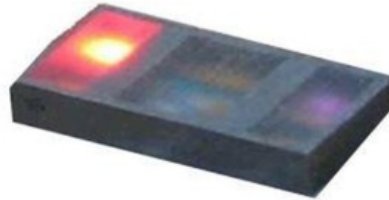


Figure 11; NJL5501R optical sensor [35].

Its features include peak wavelength of 660nm +3% tolerance (RED), and 940nm +10% tolerance (Infrared). A high output current typically of 145 μ A to 580 μ A when using infrared, and in an incredibly small package and its typical application is for pulse rate monitoring and detecting blood saturation. Its electrical characteristics can be found below.

■ ELECTRO-OPTICAL CHARACTERISTICS (Ta=25°C)						
PARAMETER	SYMBOL	TEST CONDITION	MIN	TYP	MAX	UNIT
Emitter						
Forward Voltage	V _F RED	I _F =4mA	1.7	—	2.1	V
	V _F IR	I _F =4mA	1.0	—	1.35	V
Reverse Current	IR RED	V _R =5V	—	—	5	μ A
	IR IR	V _R =5V	—	—	5	μ A
Peak Wavelength * ¹	λ_p RED	I _F =20mA	657	660	663	nm
	λ_p IR	I _F =20mA	930	940	950	nm
Detector						
Dark Current	I _{CEO}	V _{CE} =10V	—	—	0.2	μ A
Collector-Emitter Voltage	V _{CEO}	I _C =100 μ A	16	—	—	V
Coupled						
Output Current * ²	I _O RED	I _F =4mA, V _{CE} =2V, d=0.7mm	1000	—	4300	μ A
	I _O IR	I _F =4mA, V _{CE} =2V, d=0.7mm	145	—	580	μ A
Operating Dark Current * ³	I _{CEOD} RED	I _F =4mA, V _{CE} =2V	—	—	5	μ A
	I _{CEOD} IR	I _F =4mA, V _{CE} =2V	—	—	1	μ A
Response Time(Rise/Fall)	T _r , T _f	I _O =100 μ A, V _{CE} =2V, d=0.7mm, R _L =1k Ω	—	20	—	μ s

Figure 12; electrical characteristics of NJL5501R [36].

The device emits infrared light at a suitable wavelength of 940nm, although with a 10% tolerance its accuracy may provide insufficient data.

5.1.2 TCRT1000

The second choice considered for the proposed system is that of Vishay's TCRT1000 series, a reflective optical sensor with built in IR-LED and phototransistor contained in a leaded package to block visible light.

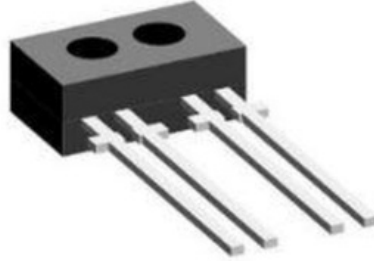


Figure 13; TCRT1000 optical sensor [37].

Another suitable sensor for the requirements with its wavelength emitting at 950nm. Other notable features include its peak operating distance at 1mm, however capable of measurements between 0.2mm and 4mm in diameter. Surrounded in a leaded package which blocks visible light, allowing for a more accurate representation by filtering unwanted artefacts on the signal caused by environmental surroundings. Its typical electrical characteristics can be found below.

BASIC CHARACTERISTICS ($T_{amb} = 25^{\circ}C$, unless otherwise specified)						
PARAMETER	TEST CONDITION	SYMBOL	MIN.	TYP.	MAX.	UNIT
SENSOR						
Collector current	$V_{CE} = 5V$, $I_F = 20mA$, $d = 1mm$ (figure 2)	$I_C^{(1)}$	0.3	0.5		mA
Cross talk current	$V_{CE} = 5V$, $I_F = 20mA$, (figure 1)	$I_{CX}^{(2)}$			1	μA
Collector emitter saturation voltage	$I_F = 20mA$, $I_C = 0.1mA$, $d = 1mm$ (figure 2)	$V_{CEsat}^{(1)}$			0.3	V
INPUT (EMITTER)						
Forward voltage	$I_F = 50mA$	V_F		1.25	1.6	V
Radiant intensity	$I_F = 50mA$, $t_p = 20ms$	I_e			7.5	mW/sr
Peak wavelength	$I_F = 100mA$	λ_p	940			nm
Virtual source diameter	Method: 63 % encircled energy	d		1.2		mm
OUTPUT (DETECTOR)						
Collector emitter voltage	$I_C = 1mA$	V_{CEO}	32			V
Emitter collector voltage	$I_E = 100\mu A$	V_{ECO}	5			V
Collector dark current	$V_{CE} = 20V$, $I_F = 0A$, $E = 0lx$	I_{CEO}			200	nA

Figure 14; electrical characteristics of TCRT1000 [38].

5.1.3 HOA1405

Another optical sensor considered is the HOA-1405 reflective optical sensor presented by Honeywell, with an IR-LED and NPN silicon phototransistor output.



Figure 15; HOA1405 optical sensor [39].

Encased side-by-side on converging optical axes in a black thermoplastic housing, the photodetector only responds to radiation from the IR-LED when a reflective object passes within its field of view. The sensor employs an IR transmissive filter to minimize the effects of visible ambient light and to provide a smooth optical face which prevents the accumulation of airborne artefact within the path of light.

To determine the most adequate sensor to obtain PPG data, the table below was constructed to highlight each transducers peripherals and allow for an accurate comparison between the devices considered.

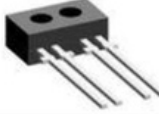
Sesnor	NJL5501R	TCRT1000	HOA-1405
			
Peripherals and characteristics	- Emitter wavelength : 660±3nm (RED) , 940±10nm (Infrared) - High output current : 1000 to 4300µA typ (RED), 145 to 580µA typ (Infrared) - Miniature, thin package: 1.9x2.6x0.8mm	- Emitter wavelength: 950 nm - Typical output current; 100µA - Dimensions (L x W x H in mm): 7 x 4 x 2.5 - Operating range; 0.2 mm to 4 mm - Lead (Pb)-free soldering released	- Emitter wavelength: 920 nm – 950 nm - Typical output current; 50 mA - Dimensions (L x W x H in mm): 51 x 25 x 2.5 - Operating range; 0.1mm to 5.5mm - IR transmissive filter to minimize the effects of visible ambient light
Cost Per Unit (£)	1.44	0.597	2.25

Figure 16; sensor comparison table.

Using the primary peripherals of each transducer listed above one sensor in particular stood out in comparison to its competition. Although each one contained an IR-LED emitting within the desired wavelength and would be somewhat appropriate for the systems application, the TCRT1000 shined in every light. Upon further investigation of the NJL5501R it became apparent that the sensor only came as a surface mount device, which would prove more challenging to implement than that of the remaining two. The HOA-1405 was significantly larger than the TCRT1000 and both of which were encased in a leaded package to minimize the effect of ambient light. However, the final decision simply came down to costing; where 3 TCRT1000s could be purchased for less than a single HOA-1405 to perform the same function. Ideally, acquiring both and testing them would have been the best approach however time constraints would prevent this.

5.2 Microcontroller Selection

The microcontroller is essentially the beating heart of any embedded system, particularly when attempting to integrate real world physical signals into a digital representation. This device contains the input/outputs (I/Os) of the system, analogue to digital converters, program memory, and often an internal oscillator. A number of microcontrollers were considered for the proposed prototype based on required specifications that are listed below. By using microchip's microcontroller selection tool it was possible to enter the systems requirements and produce a table of microcontrollers that could be a potential³.

An 8 bit microcontroller was the primary consideration for the system. Using anything above an 8 bit micro (16 bit or 32 bit) would provide an inefficient and costly method. The device would be programmed using a C compiler and would only be small in size, its variables would certainly be below that of 256 (or between -128 and 127 for signed integers). The requirements for the systems microcontroller were as follows;

- Enough input/outputs for one analogue input, an LCD in 4 bit mode (7 pins), an LED display (using 7 LEDs) with enough I/O's remaining for headroom and expansion.
- Sufficient program memory, a minimum of 14kb ideally 24kb+.
- Sufficient RAM for operation at least 256 bytes, preferably 1024+ bytes.
- Ideally an internal oscillator with an 8MHz clock frequency.

Through the use of microchips selection tool the amount of microcontrollers under consideration reduced drastically, with only three final chips remaining. The final decision was to use the PIC18F2550; the comparison table below provides justification of pic selection for the proposed system.

³ See **Appendix 3.0** for microchips controller selection tool

Product	PIC18F2550	PIC16F877	PIC16F887
Number of Pins	28	40	40
Internal Oscillator	31kHz to 8MHz	N/A	31kHz to 8MHz
Program Memory	32kb	14kb	14kb
SPI Enabled	Yes	Yes	Yes
USB Module	USB 2.0	N/A	N/A
Resolution (ADC)	12 bit	10 bit	10 bit
RAM	2048 bytes	368 bytes	368 bytes
PWM Mode	10 bit	10 bit	10 bit
Supply Voltage (V)	5.5	5.5	5.5
Operating Temp (°C)	-45 to +85	-45 to +85	45 to +85
Cost (£)	2.42	1.65	1.85

Figure 17; microcontroller comparison table.

The PIC-16 family is far from obsolete which boasts a huge number of applications where smaller, cheaper microcontrollers are more than capable of performing to standards. In particular, due to its relatively cheap costing the PIC16F series are more common within commercial applications. The comparison table above lists three primary microcontrollers and their key features; the PIC18F2550, PIC16F877 and the PIC16F887. For producing a prototype device capable of measuring PPG signals and performing calculations using a C compiler the PIC18F range provide the ideal component. Simply put, the 18F series were introduced as optimized devices for use with C programming and the use of C compilers. In fact, the release of microchips 18F series came with their very own basic free C compiler named C18. However that doesn't necessarily result in making the 16F series obsolete anytime soon.

The proposed device required an analogue input to be read via an ADC port, of which each device listed is capable of doing so to 10 bit resolution (with the exception of the 18f2550 at 12 bit). An additional 14 digital I/Os were required to implement an LCD to display BPM and HRV values to the user and a series of LED to alert the user when an input to the microcontroller is detected. All three pics listed above provide sufficient pin count for the system to work with some left over for the addition of extra features if required. In fact the 16F pics contained double the required pin

count. With such a significant amount of pins remaining the device could be implemented as a patient monitoring system, capable of measuring other health-related properties such as blood pressure, skin conductance and many others should development of the device continue.

A main feature that required consideration was the choice between using an internal or external crystal oscillator. A crystal oscillator uses an inverse piezoelectric effect; *“when an electric field is applied across a specific material it generates a mechanical deformation.”* [40]. The PIC18F2550 and 16F887 both provided internal oscillators running at up to 8MHz, whereas the 16F877 would require an external oscillator clock. Using an external clock requires purchasing additional components and therefore increasing the systems costing, however external crystals have increased accuracy and frequency stability and are of particular importance when using data communications such as USB or Ethernet. As a prototype development to prove a particular hypothesis such as this project proposes and for the particular application of the device, the increased accuracy of the external crystal is far from a necessity. Therefore the 16F877 was ruled out on this basis, however should further development of the device take place to integrate more complex features such as the use of communications for a patient monitor; the 16F877 would be a very appropriate device for consideration.

The choice between the PIC18F2550 and 16F887 was made based on a number of factors listed; program memory, RAM and an increased accuracy of ADC conversion if necessary. The key element to the decision was based upon the 18F range being designed for particular use with C compilers.

5.3 Liquid Crystal Display (LCD)

Liquid Crystal Displays (LCDs) are found everywhere, from a digital watch to television screens. They have become much more common in the television industry where the demand has increased for slimmer televisions. LCDs are incredibly slim due to their operation; millions of pixels with each pixel representing a colour i.e. red, blue, or green light that can be switched on or off very rapidly to produce a moving colour picture.

A display device was required for the system to relay information calculated via the microcontroller using the obtained PPG signals could be sent to the user. A simple 2x16 LCD was the ideal component to perform the operation as are common in many embedded systems that

incorporate the use of microcontrollers. Sparkfun's ADM1602K was selected due to the device being in stock within the laboratory at the time. However, any 2x16 LCD device would prove sufficient for the required application provided it could operate in 4-bit mode.

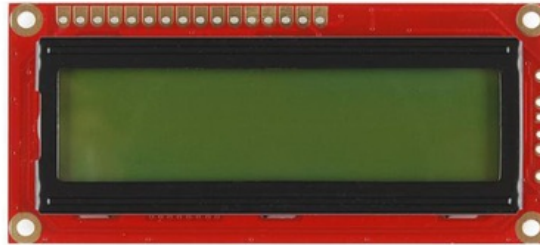


Figure 18; ADM1602K LCD [41].

5.4 Voltage Regulator

Prior to finding a suitable power source for the system it was essential to employ the use of voltage regulation. A voltage regulator generates a fixed output voltage of a particular magnitude which remains constant regardless of fluctuations within the original source of power. Linear voltage regulators employ the use of an active pass device such as a BJT or MOSFET transistor that is controlled by a high gain differential amplifier. The device compares the output voltage with a reference point and adjusts itself so that a constant output voltage is produced.

Unregulated voltage particularly when coming from a mains supply can fluctuate greatly depending on an unending number of factors. These fluctuations can often lead to lowering the life span of a system, degrading its performance, and even causing memory wipes whilst corrupting any stored data. To combat any issues regarding unstable power supply to the system a voltage regulator circuit was designed using an LM7805.

5.5 System Simulation

Circuit simulation is an important aspect of design for any system allowing for modelling of circuit operation without needing to physically build the design. Extensive circuit simulation using Proteus was done in order to determine the systems capabilities and limitations, allowing for justifiable design decisions. The simulation process was split into two key areas of the system; the signal conditioning and overall system operation.

5.5.1 Signal Conditioning

After doing the calculations to determine frequency cut-off of the RC filter and required gain of the operational amplifiers, simulation of the various proposed designs were developed through using Proteus.

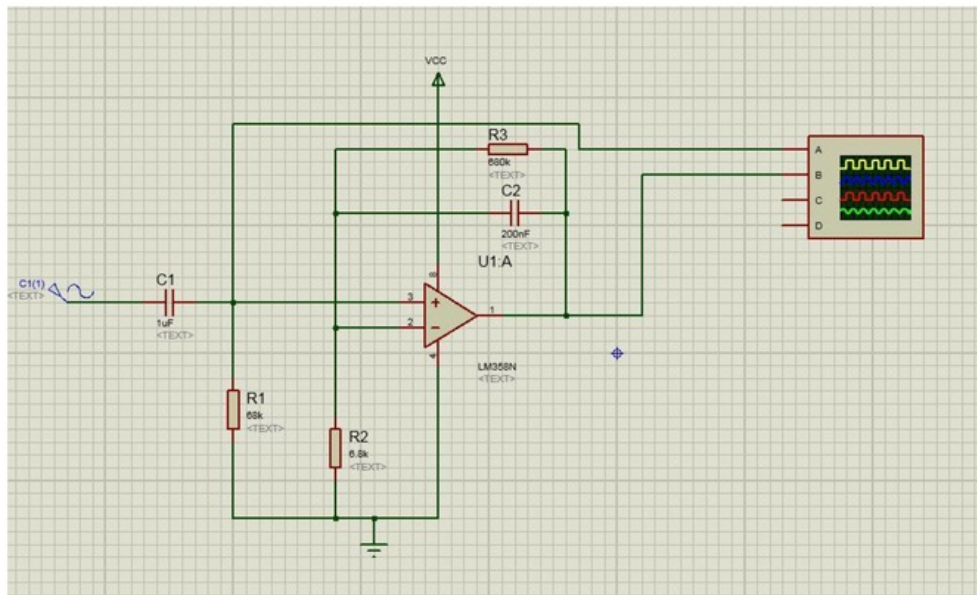


Figure 19; signal conditioning showing first stages of filtering and amplification.

The image above shows the first simulation produced in order to obtain 100 gain from one stage of amplification. Where **C2** and **R3** are the result of the RC filter and **R2** and **R3** are the resistors used to

determine the overall gain. The input and output of the circuits were measured through the use of a digital oscilloscope seen below.

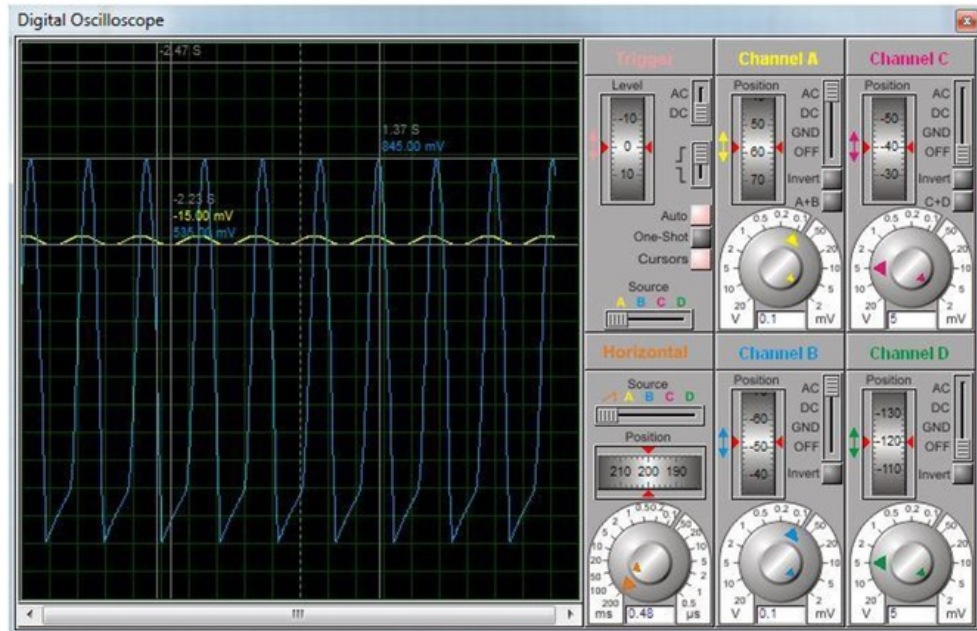


Figure 20; digital oscilloscope of first stage simulation.

The yellow trace (channel A) represents the input of the system measuring at **15mV**, the blue trace (channel B) representing the systems output measuring at **845mV**. Using the formula to determine gain the simulated circuit should provide a gain of **100**;

$$Av = \frac{680k}{6.8k} = 100$$

However, the above simulation only provides a gain of 53.3 where;

$$Av = \frac{845mV}{15mV} = 53.3$$

This was a test conducted to prove that difficulties can arise when attempting to produce a gain of **100**, where limitations are set by the operational amplifier and/or due to the magnitude of the resistors of **R3** and **R2** and impact of **C2**s internal resistance. For this reason the gain of the circuit was produced by cascading the system into two segments with reduced a reduced gain of 10 for each (as a multiple stages of 10 will provide 100 gain ($10 \times 10 = 100$)).

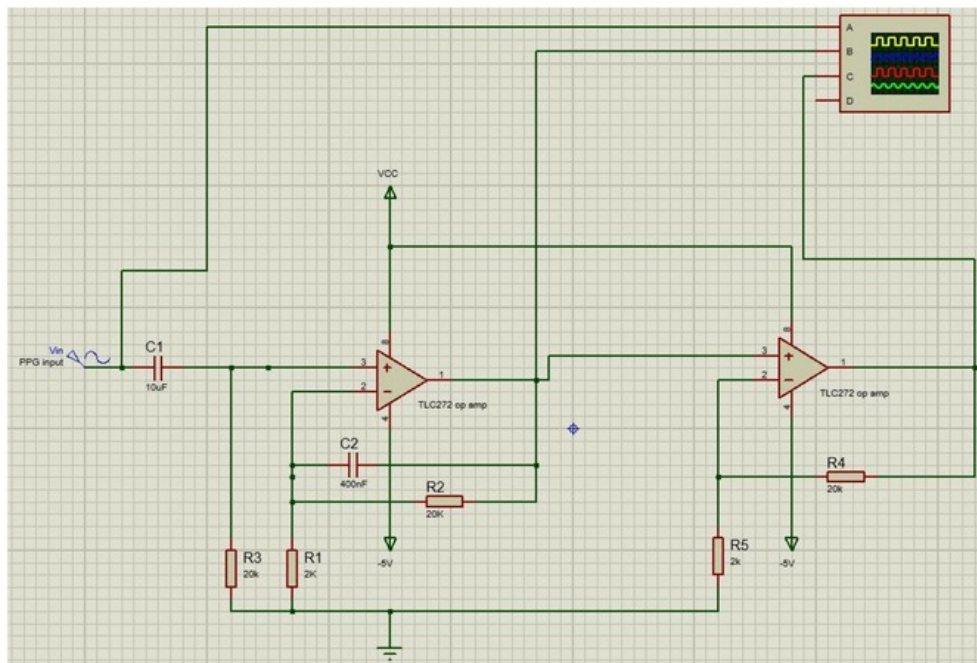


Figure 21; signal conditioning two-stage amplifier to achieve gain of 100.

By reducing the size of **R2** and **R1** from **680k/6.8k** to **20k/2k** the gain of the circuit was reduced to a magnitude of 10. The capacitor of 400nF was chosen based on the measurements undertaken in **Appendix 2.0**, where 400nF was the best trade off providing a frequency cut-off at **10Hz** with less than 0.5 loss of gain. Increasing the capacitor further to provide a better attenuation of unwanted frequencies would result in too much loss in gain.

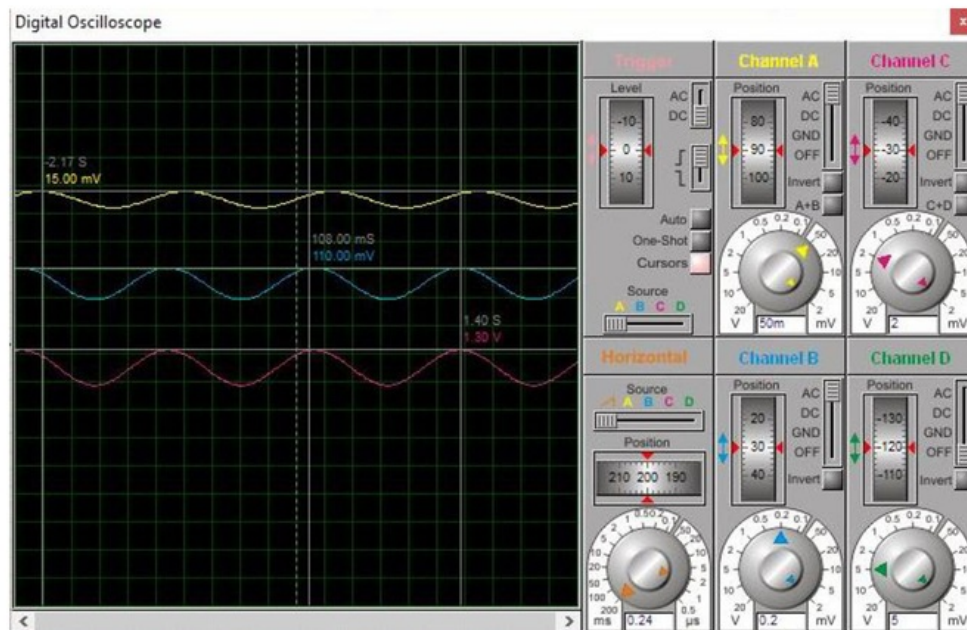


Figure 22; digital oscilloscope with 15mV input at 0.8Hz.

The yellow trace (channel A) represents the input of the circuit measuring at 15mV pk-pk, the blue trace (channel B) represents the output of the first stage of amplification at 110mV, whilst the red trace (channel C) displaying the output of the second stage at 1.30v. With the magnitude of the PPG measured at 15-30mV the design was tested under a range of various inputs to determine its validity.

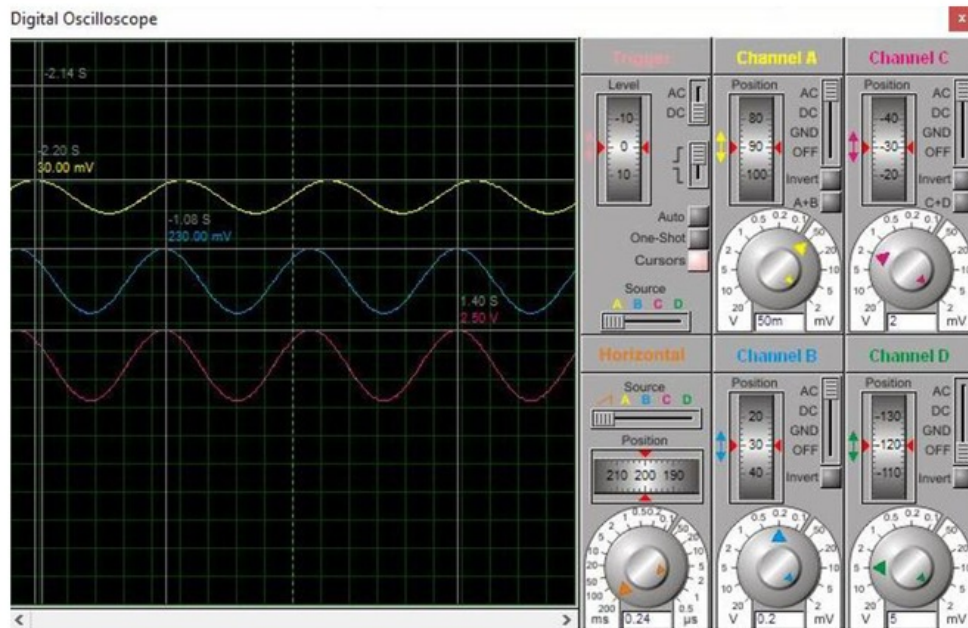


Figure 23; digital oscilloscope with 30mV pk-pk at 0.8Hz.

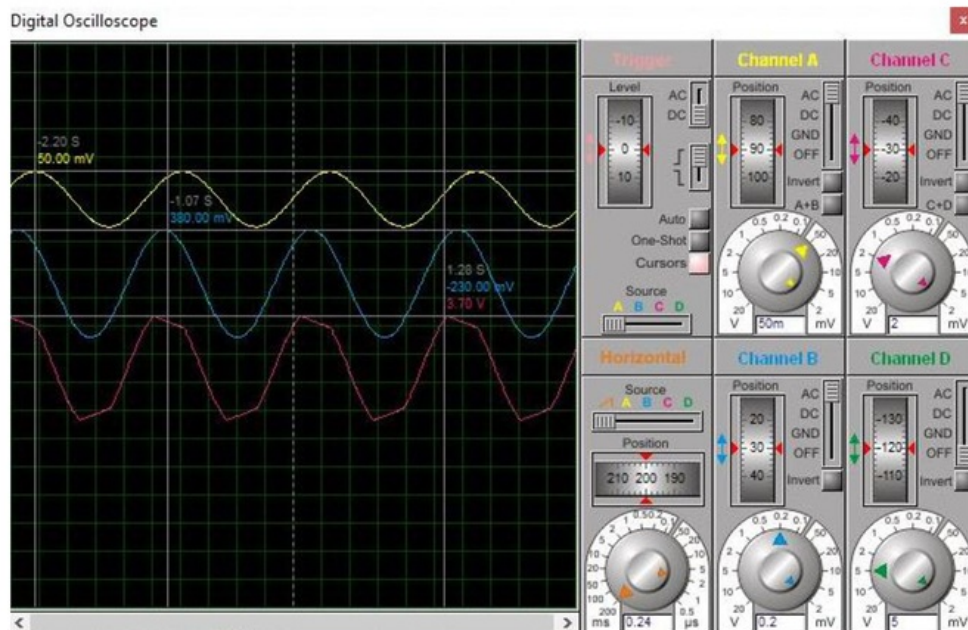


Figure 24; digital oscilloscope with 60mV pk-pk at 0.8Hz.

With an increased input signal of 50mV pk-pk the final output began to create minor distortion of the final signal. This would cause no dramatic effect on the systems functionality as the device is

only interested in calculating the number of pulses over a set period and the time intervals between them. Should an issue occur where an input pulse from the PPG sensor should exceed that of its average 15-30mV pk-pk there would be no adverse effect to the systems functionality. In fact, it would be possible to produce a square wave output to represent PPG pulses if desired distorting the final signal.

5.5.2 Microcontroller Simulation

Once the signal conditioning was sufficient the microcontroller and its outputs were simulated. By using a signal generator to represent the analogue input of the PPG signal it was possible to test the microcontroller, LEDs and LCD under various input conditions such as variations within frequency (an increase in heart rate).

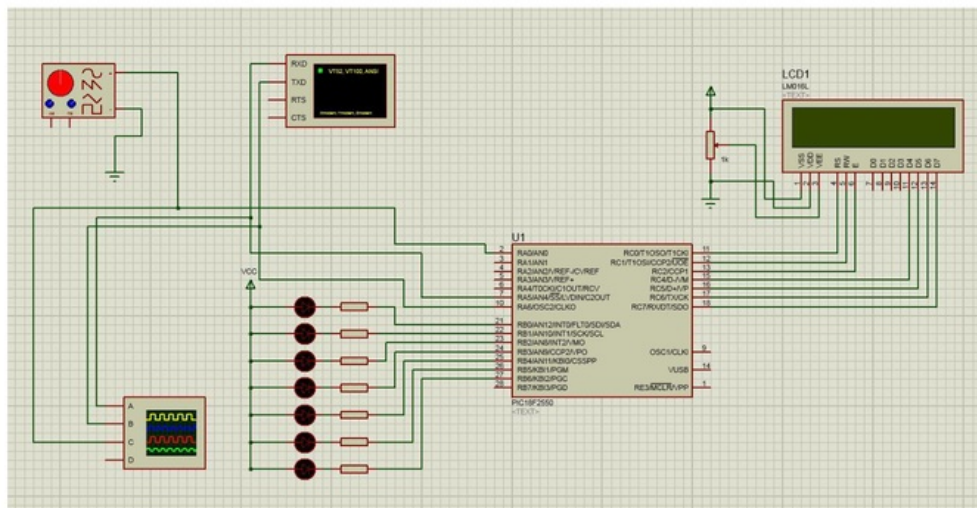


Figure 25; simulation of microcontroller, LEDs and the LCD.

The simulation above and the software implemented to test the system under varying inputs. A signal generator was connected to pin 2 (PORT AN0) of the microcontroller to represent the PPG input, with LEDs being connected from pins 21 – 27 (PORT RB0 – RB6), the LCD between pins 11-13 and 15-18 (PORT RC0-RC2, RC4-RC7). A virtual terminal was placed on pins 7 and 10 to assist in ensuring the values printed to the LCD were accurate.

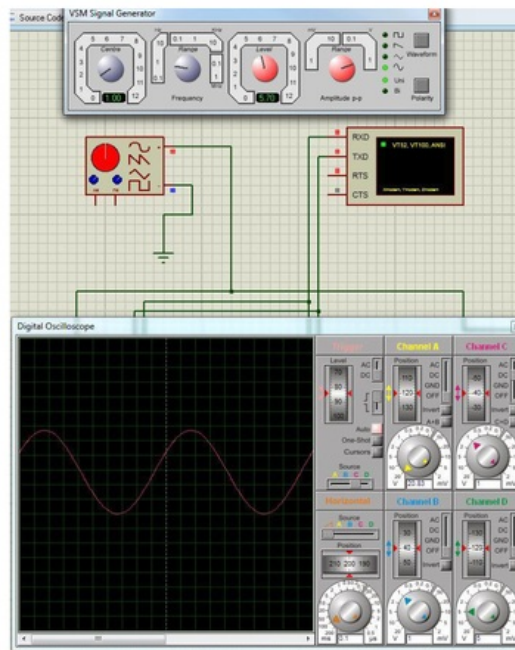


Figure 26; signal gen producing input to microcontroller of 1v at 1Hz.

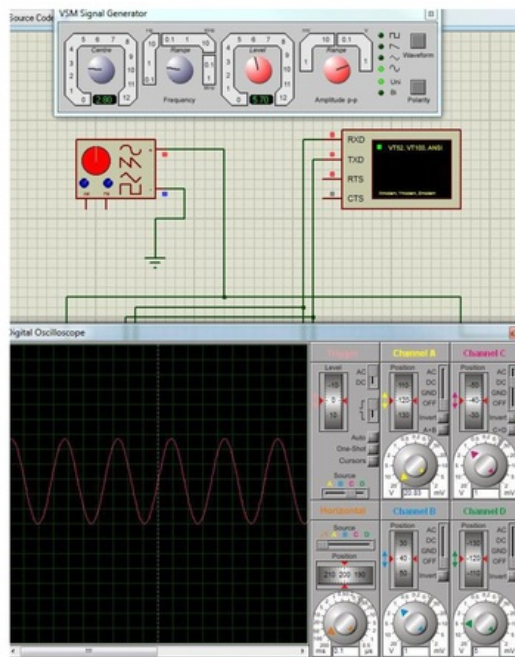


Figure 27; signal gen producing input to microcontroller of 1v at 2.8Hz.

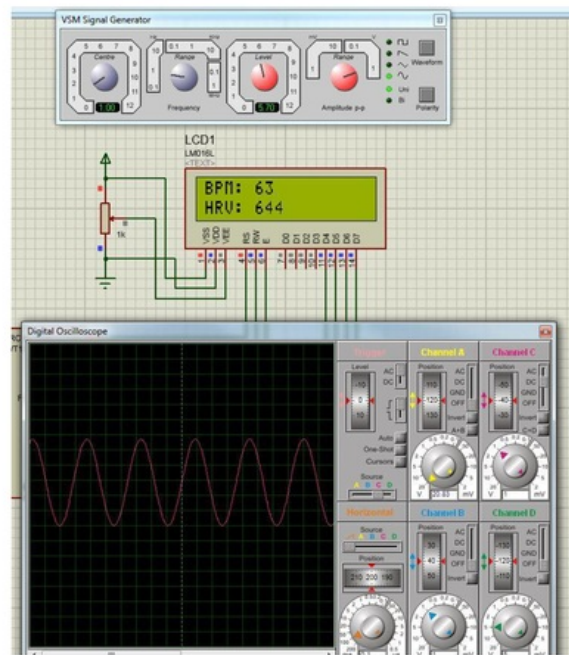


Figure 28; signal gen producing input to microcontroller of 1v at 1Hz and displaying BPM/HRV values to LCD.

Looking back at the software algorithms proposed in chapter 4, the above diagram can see is shown to be applying the algorithm through the microcontrollers' calculations. The frequency of a wave refers to how often the particles of the medium vibrate when a wave passes. In mathematical terms the frequency is the number of cycles with respect to time, known by the equation of;

$$f = \frac{1}{t}$$

Where frequencies are measured in Hertz and 1 Hertz is equivalent to 1 cycle, then the average beats per minute (BPM) of a human heart would be equivalent to 60 as shown above.

$$BPM = n * 3$$

With the microcontroller determining BPM as the number of beats over 20 second period * 3 we can see by;

$$n = \frac{63}{3}$$

The amount of beats accounted for over the 20 second period = 21, which is almost precisely equivalent to 1Hz. This test was conducted to prove the systems functionality in regards to BPM over a wide range of frequencies. These tests can be seen in the images below.

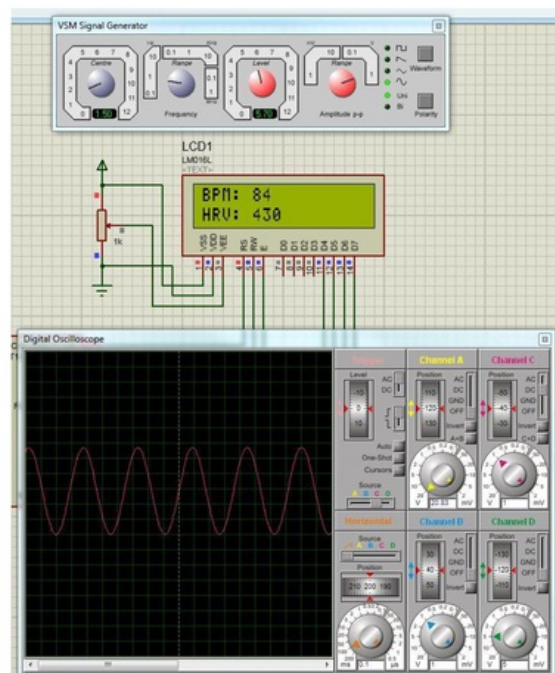


Figure 29; signal gen producing input to microcontroller of 1v at 1.5Hz and displaying BPM/HRV values to LCD.

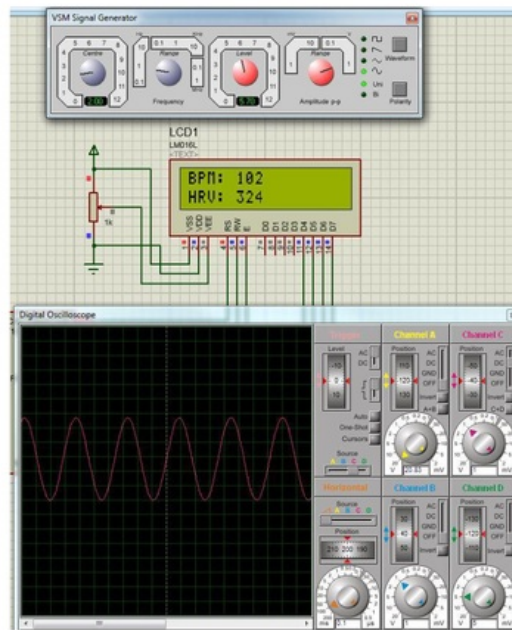


Figure 30; signal gen producing input to microcontroller of 1v at 2Hz and displaying BPM/HRV values to LCD.

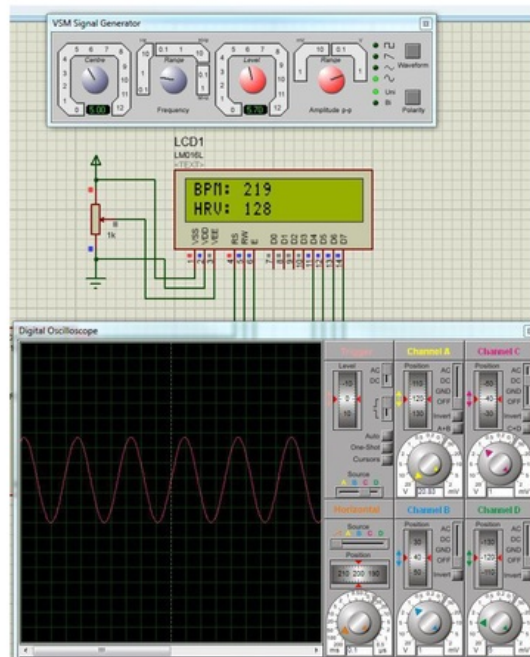


Figure 31; signal gen producing input to microcontroller of 1v at 5Hz and displaying BPM/HRV values to LCD.

The final element of the system was to provide a visual aid to the user to producing an alert when a heart rate pulse was detected via the use of LEDs. With the microcontroller using an 8 bit resolution and taking 256 samples of the PPG input, thresholds were used to determine when to set each LED to light at a specific point. I.e. the first LED within the system would display when an input PPG signal was detected and go beyond 20 samples taken.

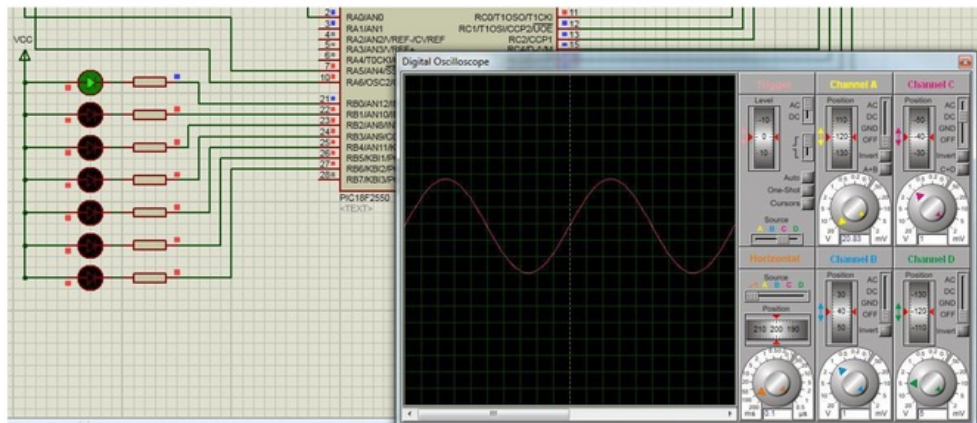


Figure 32; One LED light on when threshold set.

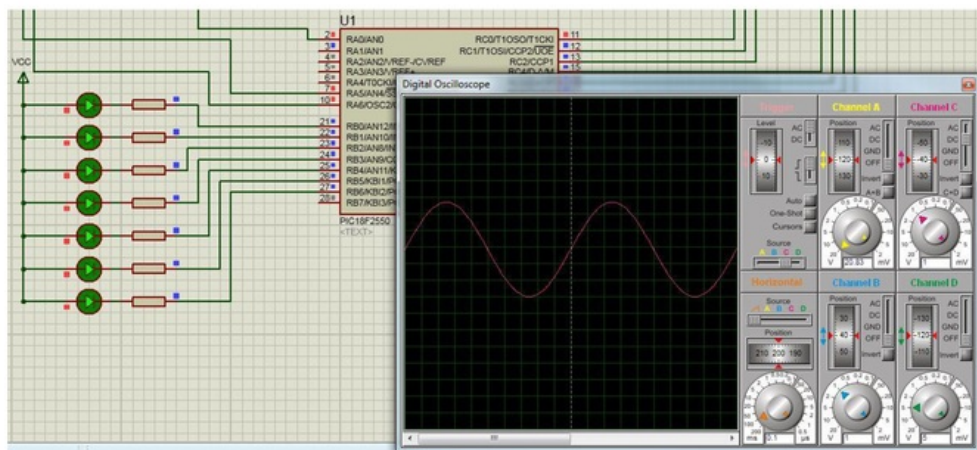


Figure 33; All LED lights on when threshold set.

With each LED having a set threshold higher than the next (20, 40, 80, 120 etc.) the system would set the series of LEDs to light in a smooth sequence of on/off when an input is detected.

5.5.3 Voltage Regulation

With the knowledge that a steady 5v supply was required for the system, the circuit for the voltage regulator was simulated in Proteus to determine an accurate steady output supply. Two 4.7uF decoupling capacitors and two 0.1uF capacitors were used to assist in producing a consistently smooth 5v output whilst removing any noise from the circuit. The circuit was simulated using a 9v supply as shown below.

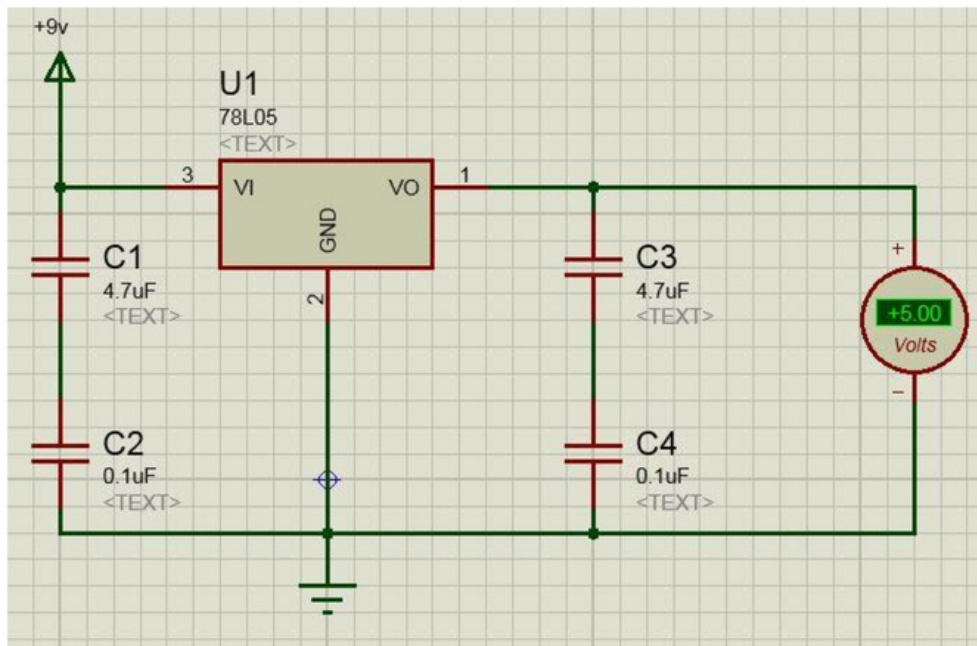


Figure 34; voltage regulator simulation.

5.6 System Costing and Commerciality

To determine whether the device could be feasible as a commercial product, an overall cost based on the components used was produced as seen in the table below.

Product	Quantity	Cost (£)
TCRT1000	1	0.60
PIC18F2550	1	2.42
LM7805	1	0.50
Welwyn MFR 20k resistor	10	0.04
Welwyn MFR 2k resistor	10	0.04
Multicomp MCRH 1uF	1	0.04
Multicomp MCGPR 470nF	1	0.04
Sparkfun ADM1602K LCD	1	17.35
Multicomp 703-0100 red LED	7	1.40
Multicomp MCKNP 220 resistor	7	1.43
Twin Industries TW-E40 Breadboard	2	15.20
Total Cost		39.06

Figure 35; costing table of prototype design

The total cost of the system using the components proposed within the systems prototype comes to £39.06. Data from the survey listed in **Appendix 6.0** highlight that 45.10% of 53 results stated they would be willing to pay £10-30 for a stress monitoring device, where 41.18% were only willing to pay less than £10. Only 11.76% of results shown would purchase a device between £30-50, therefore to produce the proposed system as a commercially viable product costing would need to be reduced to below £30 so that a larger target audience could be achieved.

Chapter 6

System Implementation

The purpose of this chapter is to highlight the approach taken to implementing the system proposed based on the designs introduced in chapter 4 and 5. This could be achieved by producing circuits on a breadboard to test the functionality of the proposed system in real-time. A number of steps were undertaken to implement the systems designs into real-time circuitry;

- Develop a voltage regulator based on the designs in chapter 5 capable of a stable 5v output to supply the circuit.
- Implementing the signal conditioning system of RC filtering and operational amplifiers to improve the quality of the PPG signal, increasing its gain whilst removing high frequency noise.
- Applying a microcontroller to the breadboard, connecting the output of the signal conditioning to the specified input of the micro.
- Create a breadboard with a series of LEDs with current limiting resistors and connecting them to the microcontroller.
- Placing an LCD to the breadboard and connecting it to the appropriate designated pins of the microcontroller.
- Implementing the software to the PIC18F2550 using the Picket 3.

Each step would involve rigorous testing to ensure the results produced were accurate to the designs of the system. The following sections of the chapter show the implementation of each step undertaken listed above.

6.1 Voltage Regulation

Before implementing any of the key components within the circuit the voltage regulation was developed so that there was no risk of spikes on the supply when testing the system. As discussed in **section 5.4** of the design chapter an **LM7805** was required to provide the system a stable and consistent supply voltage. This was achieved by applying the designs found in **section 5.5.3**, using the LM7805 with two 4.7uF de-coupling capacitors and two 0.1uF smoothing capacitors. The resulting circuit can be seen in the image below.

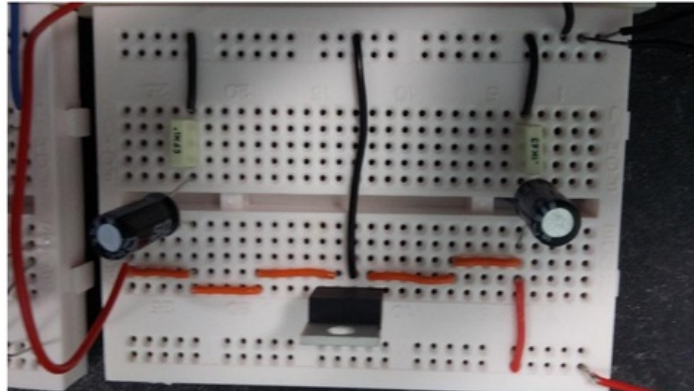


Figure 36; implementing the voltage regulator.

The device was powered using a 9v supply from the SJ Electronics function generator found within the lab and measured using a multi-meter to display its output as shown below.

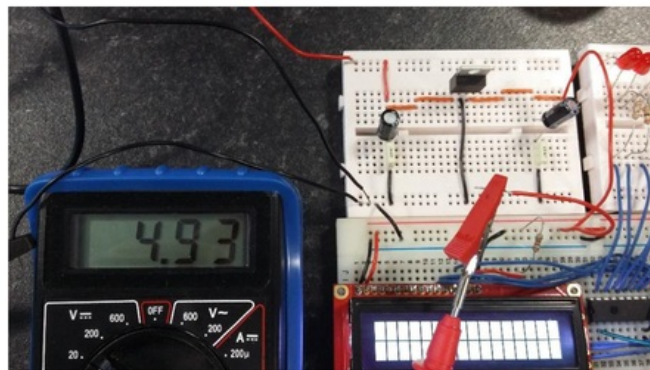


Figure 37; voltage regulator test.

The multi-meter in figure 36 shows an amplitude of 4.93v coming from the output of the regulator circuit, which is sufficiently accurate to the design proposed in **section 5.5.3**. The output of the voltage regulator circuit was connected to the power rails of the breadboard, and the ground terminal to the ground rails allowing for a steady 5v supply to the system.

6.2 Sensor and Signal Conditioning

With a steady supply voltage available when powering the system, the sensor and signal conditioning were implemented onto a separate breadboard and its signals measured and tested using a PicoScope and PicoScope 6. By applying the RC filter and operational amplifier configurations produced during the design phase it was possible to amplify the PPG signal whilst attenuating any unwanted high-frequency noise.

Figure 37 below shows the PPG sensor implementation with pin 1 and 4 are connected to the power rails between two pull-up resistors and pin 2-3 are both connected to ground. The output comes from pin 4 of the device which is then used as the input of the signal conditioning. In the image below the yellow lead is connected to the PicoScope using a probe to measure the output of the sensor pre-amplification.

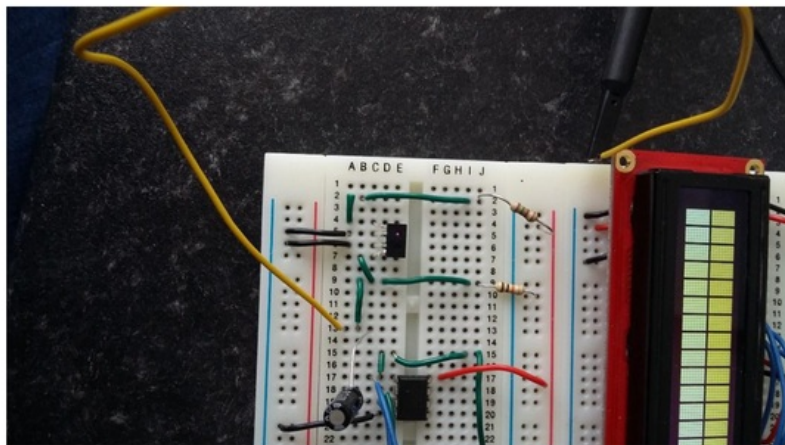


Figure 38; TCRT1000 implementation to circuit.

By testing the output of the TCRT1000 PPG sensor using PicoScope it is possible to easily identify the high-frequency artefact on the signal seen below. As seen previously in the initial designs of the system the PPG signal received pre-amplification sat at an average 10 millivolts pk-pk, which holds true in the testing show below.

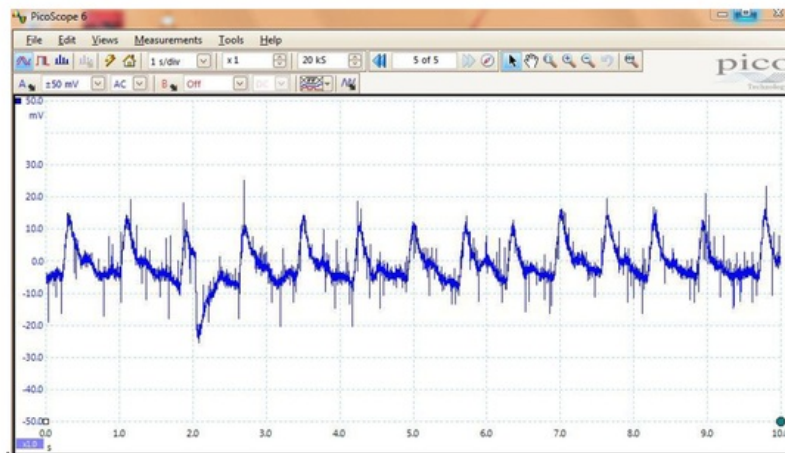


Figure 39; TCRT1000 signal pre-signal conditioning.

The following images show the measurements taken at each stage of signal conditioning; the output of stage one followed by the output of stage two.

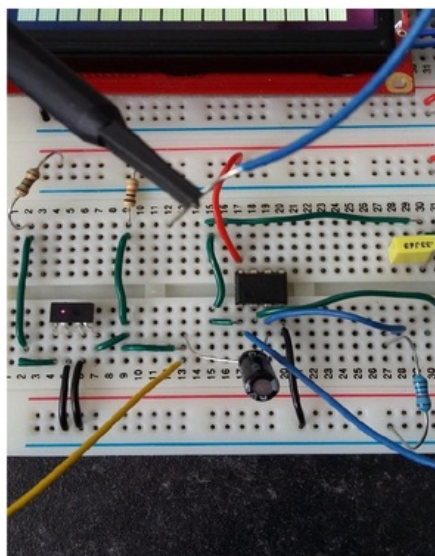


Figure 40; first stage of PPG signal amplification and filtering.

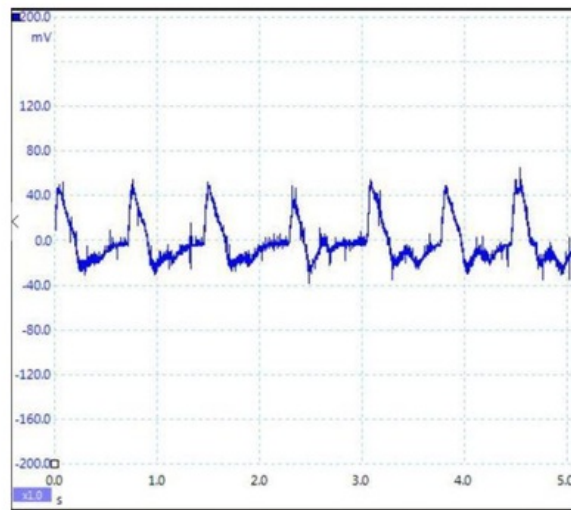


Figure 41; PPG signal output after first stage of signal conditioning.

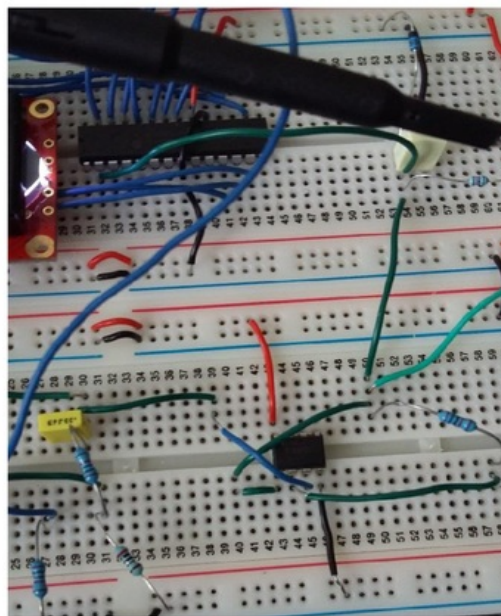


Figure 42; second stage of PPG signal amplification.

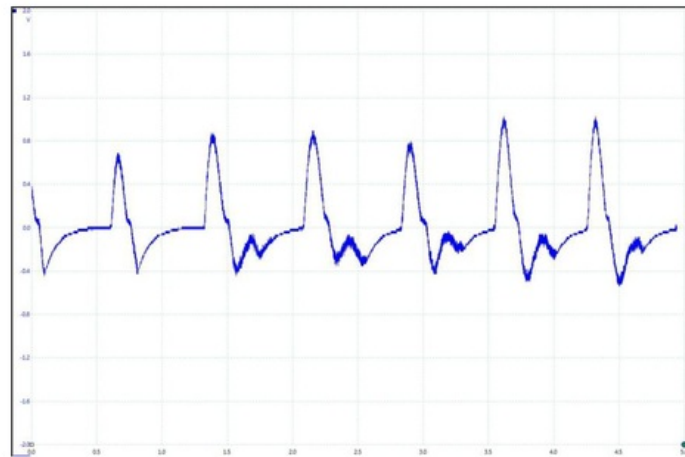


Figure 43; PPG signal output after second stage of signal conditioning.

The final output of the signal conditioning can be seen in figure 42, where the signal now sits at an average 1.2-1.4v. Increasing the gain during the first stage of amplification could be done but at the result of an increase in unwanted high-frequencies on the signal. Figure 42 shows a very clean output signal with limited noise particularly in comparison to the original signal before filtering in figure 38.

Using the built-in frequency analysis function in PicoScope it was possible to produce a graph showing at which frequency the RC filter begins to attenuate. Figure 43 shows that the RC filter begins to attenuate frequencies above 5Hz which correlates with the calculations found in **section 4.3.1** and the testing of capacitive reactance in **Appendix 2.0**.

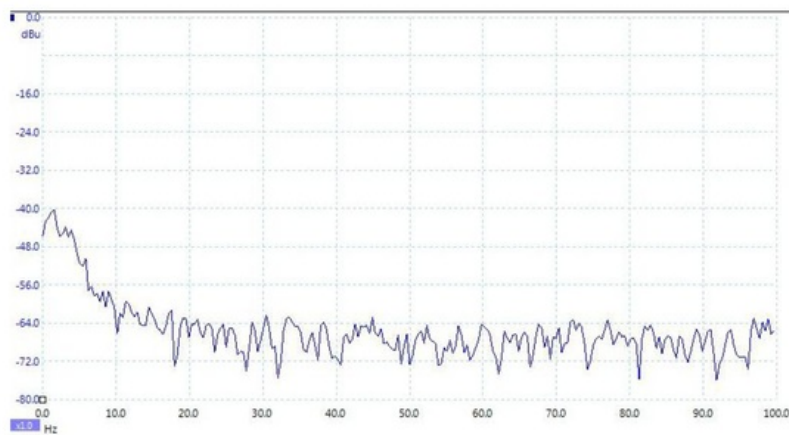


Figure 44; frequency analysis of PPG signal.

6.3 Algorithms and LCD

With a clean PPG input at a suitable amplitude being implemented into the system the software for the microcontroller was developed based on the algorithms and logic in **sections 4.2** and **4.4**. The full software listing can be found in **Appendix 4.0**, but the primary functions are discussed below. The microcontroller was configured based on the requirements of the system highlighted in **section 4.1** using the CCS C compiler, allowing for pin allocation and interrupts to be easily implemented into the microcontroller.

```
10 #include <main.h>
11 #define LCD_ENABLE_PIN PIN_C2
12 #define LCD_RS_PIN PIN_C0
13 #define LCD_RW_PIN PIN_C1
14 #define LCD_DATA4 PIN_C4
15 #define LCD_DATA5 PIN_C5
16 #define LCD_DATA6 PIN_C6
17 #define LCD_DATA7 PIN_C7
18
19 #include <lcd.c>
20 int secs=0;
21 int timer_flag=0,hrv_flag=0;
22 unsigned long int hrv_timer=0;
```

Figure 45; software definitions.

Figure 44 highlights the code used for the LCD's compatibility with the microcontroller, setting PORT C0 to C7 as the connections between the two devices. The three lines of code at the bottom show various variables being declared which are used to produce the interrupts required.

```
24 #INT_RTCC
25 void RTCC_isr(void)
26 {
27     secs++;
28     if(secs==20)
29     {
30         timer_flag=1;
31         secs=0;
32     }
33 }
34 }
```

Figure 46; software timer for 20 second interrupt.

In order to implement the algorithm for BPM to be calculated a 20 second timer was required. The timer would be used as a point of reference for the system. Using the timer as a reference point allowed the code to know when to implement the algorithm, multiplying the amount of pulses taken

by 3 and then resetting the pulse count whenever the timer was reset. This method was performed using a flag system i.e. when the timer would reach 20 seconds a flag would be set and called into the algorithm shown below so that the system would know when to calculate BPM, print to the LCD and reset the pulse count.

```

103     if(timer_flag)
104     {
105         printf("Avg bpm = %Lu\n\r", (bpm*3));
106         old_bpm=bpm*3;
107         bpm=0;
108         timer_flag=0;
109         lcd_gotoxy(1,1);
110         printf(lcd_putc, "BPM: %Lu ", old_bpm);
111     }

```

Figure 47; if statement to apply BPM algorithm.

The second timer in the system was used to determine the time taken between pulses. Another flag was implemented through the use of an if statement, incrementing the timer whenever a heartbeat or pulse dissipated and would then stop incrementing when the next pulse was measured.

```

35     #INT_TIMER2
36     void TIMER2_isr(void)
37     {
38         if(hrv_flag)
39         {
40             hrv_timer++;
41         }
42     }
43

```

Figure 48; if statement used to count time between heartbeats.

Applying the HRV algorithm in software would be slightly more complex than the simple if statement used to calculate BPM. Implementing the HRV algorithm required an array, where the time between heartbeats calculated via the timer above stored its data. Each time a new heartbeat would be detected the timer would reset to 0 so that it could accurately calculate the time from the new pulse to the next and so forth. However, the data obtained previously would be required to produce an average heart rate variability, therefore, would be stored into an array or in this case a ring counter.

```

if(hrv_timer_flag)
{
    hrv_pointer++;
    if(hrv_pointer>25)hrv_pointer=0;
    hrv_block[hrv_pointer]=hrv_timer;
    hrv_timer_flag=0;
    hrv_timer=0;
    for (a=0;a<25;a++)
    {
        hrv_avg+=hrv_block[a];
    }
    hrv_avg=hrv_avg/25;
    hrv=hrv_avg;
    hrv*=2;
    lcd_gotoxy(1,2);
    printf(lcd_putc,"HRV: %Lu mS ",hrv);
}

```

Figure 49; software to apply HRV algorithm.

A ring counter can be seen as an array in the sense that it stores multiple values into one memory location while accessible from a single variable, similar to storing data in a table. HRV block represents the ring counter where hrv pointer determines where the next value will be stored. By using this method, it was possible to store the times measured between pulses into the ring counter up to a maximum of 30 values (where hrv_block[30]). Although only 25 were required to produce an average, 30 locations were allocated to provide added headroom. The calculation shown above was then performed to provide the average variations between heartbeats in milliseconds and printed to the LCD.

6.4 Visual Aid

Developing a visual aid for the system would be an additional simple feature alerting the user when a heartbeat is detected by flashing a series of LEDs in real time. The LEDs used can be seen below;

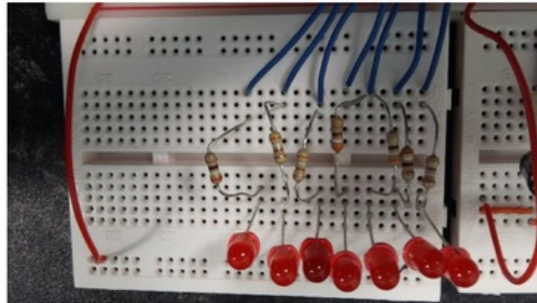


Figure 50; LED connections to breadboard.

To ensure the LEDs used were working correctly and bright enough they were first placed onto a separate breadboard with current limiting resistors and a 5v supply applied to the circuit. Multiple LEDs required replacing which took trial and error to establish a sufficient LED display, the final result can be seen below.

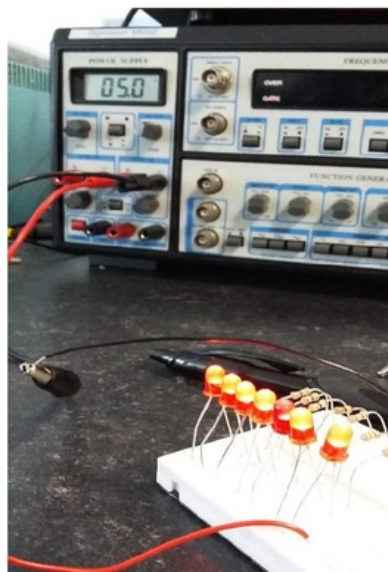


Figure 51; LED test.

For the LEDs to be used in conjunction with the system some basic software using if statements and thresholds was developed. Each LED would have a set threshold, and upon reading the analogue

input the threshold would determine when to set off which LED. The below software was implemented so that the LEDs would flash in sequence in real-time with detected heartbeats.

```
72
73     analog = read_adc();
74
75     if(analog>20) output_low(PIN_B0);
76         else output_high(PIN_B0);
77     if(analog>40) output_low(PIN_B1);
78         else output_high(PIN_B1);
79     if(analog>80) output_low(PIN_B2);
80         else output_high(PIN_B2);
81     if(analog>120) output_low(PIN_B3);
82         else output_high(PIN_B3);
83     if(analog>150) output_low(PIN_B4);
84         else output_high(PIN_B4);
85     if(analog>200) output_low(PIN_B5);
86         else output_high(PIN_B5);
87     if(analog>250) output_low(PIN_B6);
88         else output_high(PIN_B6);
```

Figure 52; setting threshold of individual LEDs.

Chapter 7

Evaluation

Listed below are the initial outcomes that were achieved throughout this project, with an evaluation of primary and secondary research, and objectives that were met to follow.

- The project was successful in producing a report containing a comprehensive review of secondary research conducted in key topic areas.
- Successful primary research with the use of a questionnaire to produce quantitative data based on the stress of final year undergraduate students and feasibility of a stress monitoring device.
- A proposed system consisting of hardware and software to produce a device capable of obtaining information regarding the hearts cardiac cycles through the use of photoelectric plethysmography.
- Sufficient signal conditioning methods applied to produce an accurate PPG signal of both amplitude and frequency.
- Implementation of successful algorithms through rigorous simulation processes using Proteus.

7.1 Evaluation of Research and Hypotheses

With extensive research into the literature of photoelectric plethysmography and heart rate variabilities relation to stress, both hypotheses proposed could be answered with a degree of success.

Can photoelectric plethysmography be used to determine stress through heart rate variability?

The initial hypothesis of the project above can be answered with almost certainty; secondary research conducted and discussed in the literature review has shown that stress can be determined based on variations of heartbeats within the time domain. This can also be seen within the market research, particularly in Polar's M400 Smart Watch HRV monitor, Olive's wristband and Android / IOS application, and Inner Balance's earlobe EKG sensor. All of which are able to determine levels of stress by measuring heart rate variability. One thing to note is that all three devices are of high cost; above £100. With some fine tuning, additional features and cost efficiency methods, the proposed system could easily be implemented as a low-end commercial product, offering a cheaper alternative for those looking to purchase a stress monitoring device.

Are final year undergraduate students under increasing stress levels to prevent the risk of under-achieving?

Due to time constraints limiting the ability to finish the production of the prototype system, no empirical data was able to be obtained in order to reinforce the research conducted into answering this final hypothesis. However, with a degree of confidence the research conducted shows that final year undergraduate students are in fact under increasing stress levels. From the data obtained from the questionnaire conducted, where over 85% of those who completed the final year of an undergraduate degree found the year incredibly stressful (answering either 4 out of 5 or 5 out of 5). This correlates directly to the BBC news report published in [42], which was one of the key driving forces of this projects topic. Personal experience also plays a role here in proving the hypothesis to a degree, where sometimes extreme pressure was found when attempting to meet all deadlines set within the final year.

7.2 Evaluation of Objectives

In **section 1.1** of this report key aims and objectives are highlighted as goals to be met. The conclusions to each objective are discussed below.

- *Produce an extensive and comprehensive literature review of related research, particularly in the field of heart rate monitoring, electrocardiography and photoelectric plethysmography and their relation to stress.*

The first primary goal set at the beginning of the project was a review of related literature, which was achieved with a strong degree of success. Major topic areas such as photoelectric plethysmography were discussed extensively with critical analysis of research conducted within the field of PPG.

- *Conduct primary research with questionnaires to obtain a consensus on stress levels of final year undergraduate students, and the marketability of a stress monitoring device.*

Quantitative data analysis was achieved with the use of both questionnaires and market research. The data received from the questionnaires was of particular use, by supporting a general consensus of final year undergraduate stress and the feasibility of a commercial product.

- *The design of a prototype portable device capable of obtaining information regarding the hearts cardiac cycles through the use of photoelectric plethysmography.*

A successful design of a prototype portable device able to measure the hearts cardiac cycle through using PPG techniques was produced in Chapters 3 and 4, with significant testing and simulation to achieve the above objective.

- Hardware and software implementation of said designs with sufficient and rigorous testing ensuring functionality of the proposed system.

Both hardware and software based on the design of the proposed system were implemented to a degree of success, however limited time resulted in the final product lacking testing to ensure the device's full functionality. Minor fault finding and problem solving would be applied to resolve any issues provided more time was set.

- Obtain empirical data from final year undergraduate students at the University of Wales Trinity Saint David in regards to their levels of stress.

This objective was not met due to the time constraints of the project.

7.2 Further Development and Recommendations

Should further development of the project be undertaken, in particular the prototype proposed the additional features and general improvement of the system could be endless. Some ideas to consider are listed below.

- To undergo fault finding and problem solving of the systems hardware implementation in order to produce a fully functional and tested system.
- Obtain empirical data relating to levels of stress of final year undergraduate students should previous recommendation be realised.
- Production of a pulse-width modulated (PWM) signal relating to heart rate data obtained by the system so interaction with a patient monitoring device or data-logging system could be achieved.
- Implementation of the PIC16F887 instead of the PIC18F2550 for the use of an external crystal oscillator to provide a more stable accuracy allowing for USB or SPI communications, displaying data obtained to a monitor such as the screen of a personal computer.
- Investigation, design and implementation of galvanic skin response (GSR) methods of obtaining heart rate data to further the systems capabilities and improve its accuracy in determining levels of stress.
- Development of a printed circuit board (PCB) design using easy-PC and applying the system to a wearable wrist device.
- Implementing the device into a plug-and-play system, capable of interacting with Android and IOS systems with software development to display levels of stress and average heart rate on an application.

Chapter 8

Conclusions

From the tests conducted of the proposed system in Chapter 6, it is clear that photoelectric plethysmography can be used to detect information regarding the heart's cardiac cycle such as average beats per minute and heart rate variability. These tests correlate strongly with the research discussed in Chapter 2 such as in [8], [9], [10] and much of the other literature reviewed.

The research philosophy of positivism was proven to be an appropriate research methodology for the project, where the proposed hypotheses discussed above were measured against a form of accepted knowledge with reinforcement from quantifiable results and analysis produced by the designed prototype.

The design and implementation processes of the project were met with a strong degree of success. Although the system was not fully integrated into a physical working prototype, the measures taken within the design and simulation prove that should more time be given to work on the project that the system could easily be successfully implemented. Should the device be fully operational then empirical data could be collected to assist in answering the second hypothesis with a stronger degree of accuracy.

The project undertaken was met with strong success, and also with failures. With a stronger understanding of signal conditioning techniques, the system could have been implemented within the time given. However the overall design of the system was met, with a comprehensive report on relative research, and highlighting key design and implementation steps produced.

Chapter 9

Reflection

The work undertaken throughout the project has resulted in a great deal of personal development. It was academically and intellectually challenging but a surprisingly enjoyable experience instead of a long and daunting task that was originally expected.

Throughout all stages of the project large amounts of knowledge have been gained in a number of interesting topics. In particular in analogue signal conditioning and C programming where basic foundations were already present but have since increased dramatically in their understanding of basic and more complex concepts.

This project provided an introduction into new interesting topic areas, particularly that of optical electronics with the use of infrared light to obtain information. Optical engineering is definitely a topic area that will be considered in future endeavours if and where possible.

Although comfortable with skills in undertaking individual research and producing reports in an academic format, the project undertaken has had a great deal of impact on both qualities and an improvement in written communication skills. As well as proving how important time management can be when undertaking such a significant amount of work.

The use of the hardware and software applications listed in **section 3.2** have also seen drastic improvements. Before conducting the project soldering skills in particular were of little to no finesse, however after multiple attempts of soldering the connections for the LCD those skills have developed considerably.

To conclude, the project was challenging and yet contained a large degree of personal fulfilment. Although issues with time constraints impacted the final product of the system and its full implementation into hardware, a strong feeling of accomplishment has been achieved. The use of optical engineering techniques used throughout the project has potential to be further developed when undertaking research through the upcoming masters' year, particularly with exciting prospects in the field of renewable energy. Only time will tell.

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Appendices

Appendix 1.0 Statistical Analysis of Stress in the UK

Information in figures below obtained from data shown in [1] by the Health and Safety Executive Work.

Type of illness	Estimated prevalence (thousands)			Rate per 100 000 employed in last 12 months		
	central	95% C.I.		central	95% C.I.	
		lower	upper		lower	upper
Musculoskeletal disorders	553	510	595	1740	1600	1870
mainly affecting the upper limbs or neck	233	206	260	730	650	820
mainly affecting the lower limbs	97	80	114	310	250	360
mainly affecting the back	223	195	250	700	610	780
Breathing or lung problems	37	26	48	120	81	150
Skin problems	20	12	27	62	37	86
Hearing problems	14	7	20	42	23	62
Stress, depression or anxiety	440	403	477	1380	1270	1500
Headache and/or eyestrain	25	16	35	80	50	110
Heart disease/attack, other circulatory system	18	10	27	57	31	84
Infectious disease (virus, bacteria)	25	16	33	78	52	100
Other type of illness	110	92	129	350	290	400
Total	1243	1180	1307	3910	3710	4100

Figure 53; "estimated prevalence and rates of self-reported illness caused or made worse by work, by type of illness, for people working in the last 12 months, 2014/15" [1].

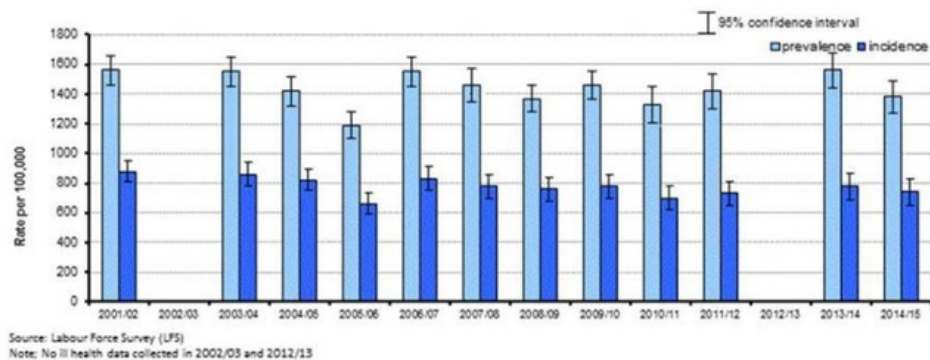
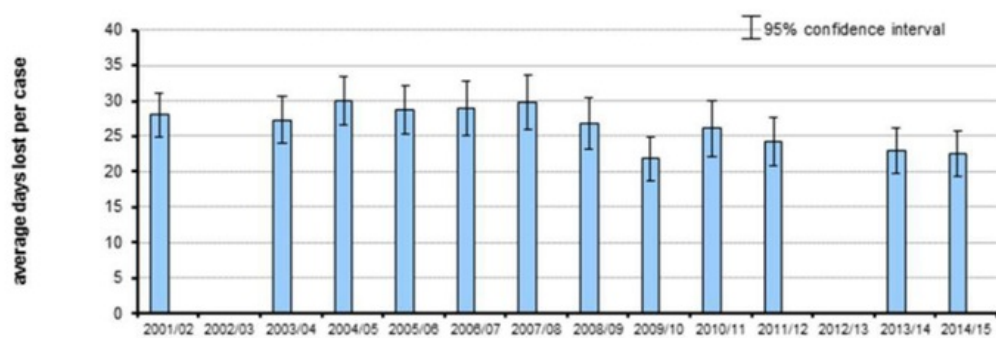


Figure 54; "estimated prevalence and incidence rates of stress, depression or anxiety caused or made worse by work, for people working in the last 12 months" [1].



Source: Labour Force Survey
 Note: No ill health data was collected in 2002/03 or 2012/13

Figure 55; "estimated average days lost per case due to work related stress, depression or anxiety injury, for people working in the last 12 months, 2014/15" [1].

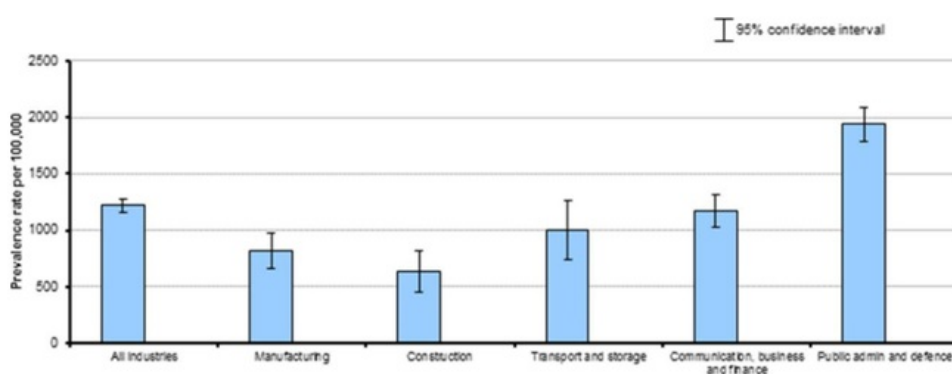


Figure 56; stress induced work loss per industry [1].

Appendix 2.0 Capacitive Reactance

Capacitive reactance calculations and tests done by implementing basic theory into an excel spreadsheet and applying the formula of;

$$X_c = \frac{1}{2\pi fC}$$

The above formula is applied into block **G6** by using the following; $1/(2 * PI() * C6 * H52)$. Where **C6** = frequency of the signal, and **H52** represents block **H2** where the capacitance can be changed freely to calculate the values of **Xc** over a range of frequencies. Column J represents the total resistance of **R2 (20k)** in parallel with the capacitive reactance of **Xc** which is given by;

$$R_t = \frac{R_2 * X_c}{R_2 + X_c} = \frac{2000 * X_c}{2000 + X_c}$$

Where **Rt** represents total resistance of both components in parallel. This will in turn effect the overall gain of the system, which is the relationship between determining a suitable capacitor that does not generate a significant loss in gain but is capable of attenuating high frequencies at a suitable frequency cut-off.

	A	B	C	D	E	F	G	H	I	J	K	L
1												
2	r1	2000		r2	20000		C2	0.0000004				
3												
4			Frequency (Hz)		R2		Xc			Total Resistance		Gain
5												
6			1		20000		397887.4			19042.80426		9.521402

Figure 57; capacitive reactance proposed formula using Microsoft Excel 2013.

The method proposed above was used in order to determine a suitable value of capacitor for **C2** by testing this theory over a range of frequencies. The final result was **400nF** as seen above as **0.0000004** which provided the least amount of gain loss with a reasonable **Fc**. Images of applying this method using a range of capacitors from **100nF** to **4.7uF** can be seen below.

31		20000	51340.3		14393.07129	7.196536
32		20000	49735.92		14264.07508	7.132038
33		20000	48228.77		14137.37055	7.068685
34		20000	46810.28		14012.89718	7.006449
35		20000	45472.84		13890.59655	6.945298
36		20000	44209.71		13770.41226	6.885206
37		20000	43014.85		13652.28984	6.826145

Figure 58; excel spreadsheet to determine circuit characteristics with a 100nF capacitor.

14		20000	56841.05		14794.44914	7.397225
15		20000	53051.65		14524.42193	7.262211
16		20000	49735.92		14264.07508	7.132038
17		20000	46810.28		14012.89718	7.006449
18		20000	44209.71		13770.41226	6.885206
19		20000	41882.88		13536.1767	6.768088
20		20000	39788.74		13309.77659	6.654888

Figure 59; excel spreadsheet to determine circuit characteristics with a 200nF capacitor.

11		20000	48228.77		14137.37055	7.068685
12		20000	44209.71		13770.41226	6.885206
13		20000	40808.96		13422.02199	6.711011
14		20000	37894.03		13090.82522	6.545413

Figure 60; excel spreadsheet to determine circuit characteristics with a 300nF capacitor.

	A	B	C	D	E	F	G	H	I	J	K	L	M	N
1														
2	r1	2000		r2	20000	C2	0.0000004							
3														
4			Frequency (Hz)		R2		Xc			Total Resistance		Gain		.707 of gain = Fc
5														
6			1		20000		397887.4			19042.80426		9.521402		6.731631305
7			2		20000		198943.7			18173.04614		9.086523		
8			3		20000		132629.1			17379.26811		8.689634		
9			4		20000		99471.84			16651.93068		8.325965		
10			5		20000		79577.47			15983.02715		7.991514		
11			6		20000		66314.56			15365.78763		7.682894		
12			7		20000		56841.05			14794.44914		7.397225		
13			8		20000		49735.92			14264.07508		7.132038		
14			9		20000		44209.71			13770.41226		6.885206		
15			10		20000		39788.74			13309.77659		6.654888		

Figure 61; excel spreadsheet to determine circuit characteristics with a 400nF capacitor.

	A	B	C	D	E	F	G	H	I	J	K	L	M	N
1														
2	r1	2000		r2	20000	C2	0.0000005							
3														
4			Frequency (Hz)		R2		Xc			Total Resistance		Gain		.707 of gain = Fc
5														
6			1		20000		318309.9			18817.65205		9.408826		6.65204
7			2		20000		159154.9			17767.29577		8.883648		
8			3		20000		106103.3			16827.99725		8.413999		
9			4		20000		79577.47			15983.02715		7.991514		
10			5		20000		63661.98			15218.85553		7.609428		
11			6		20000		53051.65			14524.42193		7.262211		
12			7		20000		45472.84			13890.59655		6.945298		
13			8		20000		39788.74			13309.77659		6.654888		
14			9		20000		35367.77			12775.57982		6.38779		

Figure 62; excel spreadsheet to determine circuit characteristics with a 500nF capacitor.

	A	B	C	D	E	F	G	H	I	J	K	L	M	N
1														
2	r1	2000		r2	20000	C2	0.0000006							
3														
4			Frequency (Hz)		R2		Xc			Total Resistance		Gain		.707 of gain = Fc
5														
6			1		20000		265258.2			18597.7618		9.298881		6.574308795
7			2		20000		132629.1			17379.26811		8.689634		
8			3		20000		88419.41			16310.62381		8.155312		
9			4		20000		66314.56			15365.78763		7.682894		
10			5		20000		53051.65			14524.42193		7.262211		
11			6		20000		44209.71			13770.41226		6.885206		
12			7		20000		37894.03			13090.82522		6.545413		

Figure 63; excel spreadsheet to determine circuit characteristics with a 600nF capacitor.

	A	B	C	D	E	F	G	H	I	J	K	L	M	N
1														
2	r1	2000		r2	20000	C2	0.0000007							
3														
4			Frequency (Hz)		R2		Xc			Total Resistance		Gain		.707 of gain = Fc
5														
6			1		20000		227364.2			18382.95116		9.191476		6.498373235
7			2		20000		113682.1			17007.82683		8.503913		
8			3		20000		75788.07			15824.11455		7.912057		
9			4		20000		56841.05			14794.44914		7.397225		
10			5		20000		45472.84			13890.59655		6.945298		
11			6		20000		37894.03			13090.82522		6.545413		
12			7		20000		32480.6			12378.13601		6.189068		

Figure 64; excel spreadsheet to determine circuit characteristics with a 700nF capacitor.

	A	B	C	D	E	F	G	H	I	J	K	L	M	N
1														
2	r1	2000		r2	20000	C2	0.0000008							
3														
4			Frequency (Hz)		R2		Xc			Total Resistance		Gain		.707 of gain = Fc
5														
6			1		20000		198943.7			18173.04614		9.086523		6.424171809
7			2		20000		99471.84			16651.93068		8.325965		
8			3		20000		66314.56			15365.78763		7.682894		
9			4		20000		49735.92			14264.07508		7.132038		
10			5		20000		39788.74			13309.77659		6.654888		
11			6		20000		33157.28			12475.16048		6.23758		

Figure 65; excel spreadsheet to determine circuit characteristics with an 800nF capacitor.

	A	B	C	D	E	F	G	H	I	J	K	L	M	N
1														
2	r1	2000		r2	20000	C2	0.0000009							
3														
4			Frequency (Hz)		R2		Xc			Total Resistance		Gain		.707 of gain = Fc
5														
6			1		20000		176838.8			17967.88058		8.98394		6.351645785
7			2		20000		88419.41			16310.62381		8.155312		
8			3		20000		58946.28			14933.26317		7.466632		
9			4		20000		44209.71			13770.41226		6.885206		
10			5		20000		35367.77			12775.57982		6.38779		
11			6		20000		29473.14			11914.80429		5.957402		

Figure 66; excel spreadsheet to determine circuit characteristics with a 900nF capacitor.

	A	B	C	D	E	F	G	H	I	J	K	L	M	N
1														
2	r1	2000		r2	20000	C2	0.0000047							
3														
4			Frequency (Hz)		R2		Xc			Total Resistance		Gain		.707 of gain = Fc
5														
6			1		20000		33862.75			12573.71799		6.286859		4.444809309
7			2		20000		16931.38			9169.101363		4.584551		
8			3		20000		11287.58			7215.376166		3.607688		

Figure 67; excel spreadsheet to determine circuit characteristics with a 4.7uF capacitor.

Appendix 3.0 Microchips Selection Tool

The images below are an example of microchips selection process for assistance in finding a suitable microcontroller given a particular set of requirements.

Product	SK Pricing	Family/Core	Program Memory Size (Kbytes)	Program Memory Size (KWords)	RAM (bytes)	I/O Pins	Pin count	Max. CPU Speed MHz	Brown Out Reset (BOR)	Low Voltage Detection	Power On Reset (POR)	Internal Oscillator
PIC18F2510	\$2.97	High Perf. 8-Bit	32	16	1536	25	28	40	Programmable BOR	Programmable LVD	Yes	8 MHz, 32 kHz
PIC18F2520	\$3.18	High Perf. 8-Bit	32	16	1536	25	28	40	Programmable BOR	Programmable LVD	Yes	8 MHz, 32 kHz
PIC18F2455	\$3.30	High Perf. 8-Bit	24	12	2048	24	28	48	Programmable BOR	Programmable LVD	Yes	8 MHz, 32 kHz
PIC18F2550	\$3.44	High Perf. 8-Bit	32	16	2048	24	28	48	Programmable BOR	Programmable LVD	Yes	8 MHz, 32 kHz
PIC18F2523	\$3.46	High Perf. 8-Bit	32	16	1536	25	28	40	Programmable BOR	Programmable LVD	Yes	8 MHz, 32 kHz
PIC18F2458	\$3.58	High Perf. 8-Bit	24	12	2048	24	28	48	Programmable BOR	Programmable LVD	Yes	8 MHz, 32 kHz
PIC18F2553	\$4.12	High Perf. 8-Bit	32	16	2048	24	28	48	Programmable BOR	Programmable LVD	Yes	8 MHz, 32 kHz
PIC18F2580	\$4.17	High Perf. 8-Bit	32	16	1536	25	28	40	Programmable BOR	Programmable LVD	Yes	8 MHz, 32 kHz

Figure 68; microchip selection tool.

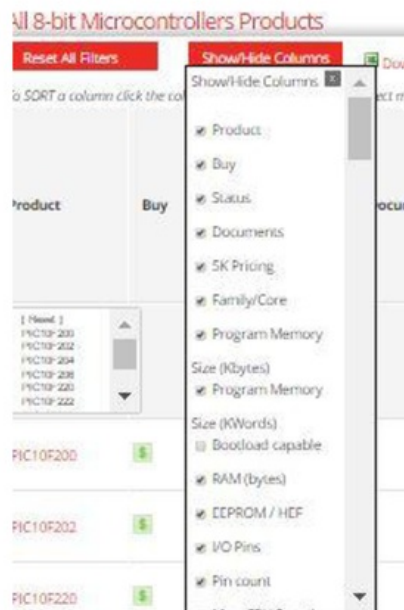


Figure 69; microchip selection tool filters.

Appendix 4.0 Software Listing

```
// File; HRmonitor (proteus project)
// Author; Matt Hunt P130507
// Date; 15/04/2016
// Description; Software for PIC18F2550 using analogue input on port A0 which calculates average
// heart rate (BPM) and determines heart rate variability. Displays data to 4 bit LCD through ports
// RC0-RC7 and a series of LED lights to show user when an input / pulse is being detected.

//-----

#include <main.h>
#define LCD_ENABLE_PIN PIN_C2
#define LCD_RS_PIN PIN_C0
#define LCD_RW_PIN PIN_C1
#define LCD_DATA4 PIN_C4
#define LCD_DATA5 PIN_C5
#define LCD_DATA6 PIN_C6
#define LCD_DATA7 PIN_C7

#include <lcd.c>
int secs=0;
int timer_flag=0,hrv_flag=0;
unsigned long int hrv_timer=0;

#INT_RTCC
void RTCC_isr(void)
{
    {
        secs++;
        if(secs==20)
        {
            timer_flag=1;
            secs=0;
        }
    }

#INT_TIMER2
void TIMER2_isr(void)
{
    if(hrv_flag)
    {
        hrv_timer++;
    }
}

Void main()
{
```

```

Unsigned int analog,analog_old;
unsigned long int bpm,old_bpm=0;
Unsigned int old_hrv=0,hrv_pointer=0,hrv_timer_flag=0;
unsigned long int hrv_block[30],hrv;
float hrv_avg=0;
char a,message[10];
setup_adc_ports(AN0);
setup_adc(ADC_CLOCK_INTERNAL|ADC_TAD_MUL_0);
set_adc_channel(0);
delay_us(30);
setup_timer_0(RTCC_INTERNAL|RTCC_DIV_32);
output_low(PIN_A1);
setup_timer_2(T2_DIV_BY_1,249,16);

enable_interrupts(INT_RTCC);
enable_interrupts(INT_TIMER2);
enable_interrupts(GLOBAL);
  lcd_init();

while(TRUE)
{
  analog=read_adc();

  if(analog>20) output_low(PIN_B0);
    else output_high(PIN_B0);
  if(analog>40) output_low(PIN_B1);
    else output_high(PIN_B1);
  if(analog>80) output_low(PIN_B2);
    else output_high(PIN_B2);
  if(analog>120) output_low(PIN_B3);
    else output_high(PIN_B3);
  if(analog>150) output_low(PIN_B4);
    else output_high(PIN_B4);
  if(analog>200) output_low(PIN_B5);
    else output_high(PIN_B5);
  if(analog>250) output_low(PIN_B6);
    else output_high(PIN_B6);

  a=0;
  if ((analog >= 200) & (analog_old < 200))
  {
    bpm++;
    hrv_flag=0;
    hrv_timer_flag=1;
  }
}

```

```

if(timer_flag)
{
    printf "Avg bpm = %Lu\n\r",(bpm*3));
    old_bpm=bpm*3;
    bpm=0;
    timer_flag=0;
    lcd_gotoxy(1,1);
    printf(lcd_putc,"BPM: %Lu ",old_bpm);
}

if(analog==200)
{
    a++;
    a++;
}

if ((analog < 200) & (analog_old >= 200))
{
    hrv_flag=1;
}
analog_old=analog;
if(hrv_timer_flag)
{
    hrv_pointer++;
    if(hrv_pointer>25)hrv_pointer=0;
    hrv_block[hrv_pointer]=hrv_timer;
    hrv_timer_flag=0;
    hrv_timer=0;
    for (a=0;a<25;a++)
    {
        hrv_avg+=hrv_block[a];
    }
    hrv_avg=hrv_avg/25;
    hrv=hrv_avg;
    hrv*=2;
    lcd_gotoxy(1,2);
    printf(lcd_putc,"HRV: %Lu ",hrv);
}
}
}

```

Appendix 5.0 Market Research

Inner Balance earlobe EKG sensor with Android / IOS application

<http://www.heartmath.com/innerbalance/>

Olive wristband and Android / IOS application

<https://www.indiegogo.com/projects/olive-a-wearable-to-manage-stress#/>

H7 wristband with Android / IOS application

http://www.polar.com/uk-en/products/accessories/H7_heart_rate_sensor

FitBit Blaze Smart Watch

https://www.amazon.co.uk/Fitbit-Blaze-Smart-Fitness-Silver/dp/B019VM3CPW?ie=UTF8&keywords=fitbit&qid=1460974741&ref=sr_1_4&sr=8-4

D21 Bluetooth Bracelet

http://www.gearbest.com/smart-watches/pp_335977.html

GOW Heart Monitor Vest

<https://www.runnerinn.com/running/gow-w-o-hear-monitor/66015/p>

Polarm400 Smart Watch HRV monitor

<http://www.wiggle.co.uk/polar-m400/>

Forerunner 620 Smart Watch

<https://buy.garmin.com/en-GB/GB/sports-recreation/forerunner-620/prod122785.html>

JayBird Reign Activity Tracker (wristband)

<http://www.jaybirdsport.com/reign-active-recovery-band/>

Zensorium Tinke Activity Tracker Android/IOS application

<https://www.zensorium.com/tinke>

Cloud Watch N7 Smart Watch for Android / IOS devices

http://www.lightinthebox.com/cloud-watch-n7-wearables-smart-watch-hands-free-calls-camera-control-heart-rate-monitoring-for-android-ios-phones_p3611324.html

Samsung Gear Fit wristband

<http://www.samsung.com/uk/consumer/mobile-devices/wearables/gear/SM-R3500ZKABTU>

Withings Pulse Ox wristband

<http://www.withings.com/uk/en/products/pulse>

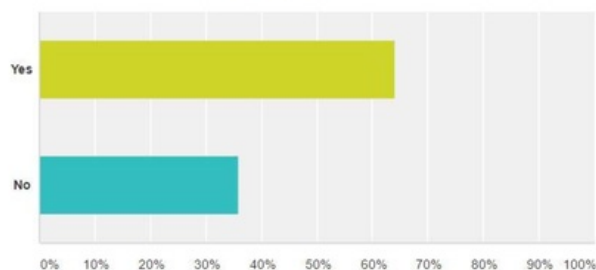
Basis Peak Fitness and Sleep Tracker

<http://www.mybasis.com/en-GB/>

Appendix 6.0 Survey Results

Do you, or have you studied the final year of an undergraduate honors degree?

Answered: 53 Skipped: 0

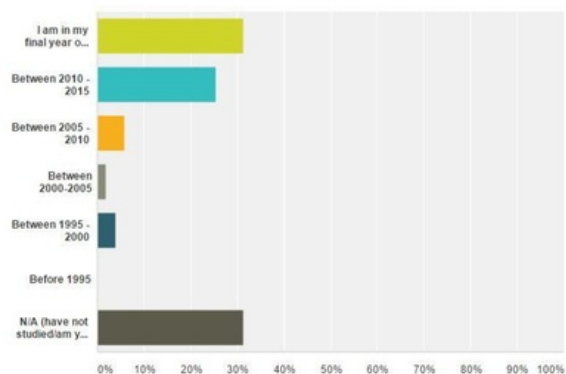


Answer Choices	Responses
Yes	64.15% 34
No	35.85% 19
Total	53

Figure 70; survey results 1.

If yes, when did you undertake the final year of study?

Answered: 51 Skipped: 2

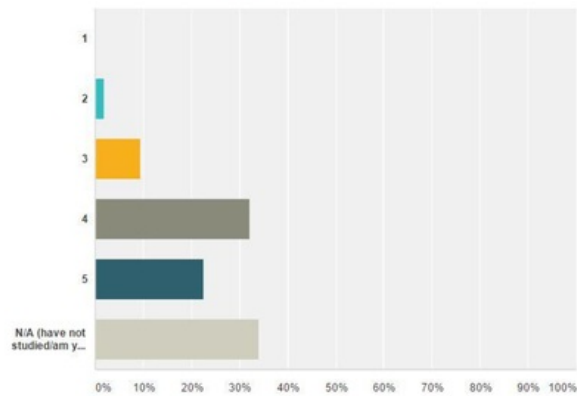


Answer Choices	Responses
I am in my final year of study	31.37% 16
Between 2010 - 2015	25.49% 13
Between 2005 - 2010	5.88% 3
Between 2000-2005	1.96% 1
Between 1995 - 2000	3.92% 2
Before 1995	0.00% 0
N/A (have not studied/am yet to study)	31.37% 16
Total	51

Figure 71; survey results 2.

On a scale of 1-5 (where 5 is the highest),
how stressful did you find your final year?

Answered: 53 Skipped: 0

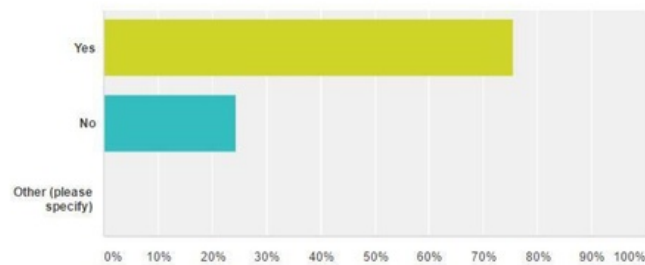


Answer Choices	Responses
1	0.00% 0
2	1.89% 1
3	9.43% 5
4	32.08% 17
5	22.64% 12
N/A (have not studied iam yet to study)	33.96% 18
Total	53

Figure 72; survey results 3.

Do you find that you work better under a
degree of stress?

Answered: 53 Skipped: 0



Answer Choices	Responses
Yes	75.47% 40
No	24.53% 13
Other (please specify)	0.00% 0
Total	53

Figure 73; survey results 4.

Have you ever used and/or considered the use of a stress monitoring device?

Answered: 53 Skipped: 0

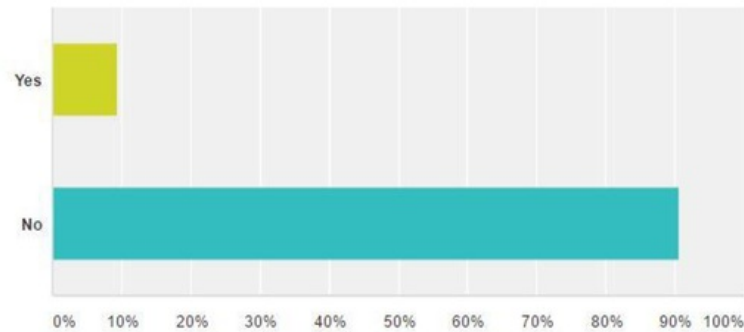


Figure 74; survey results 5.

If you were looking for such a device, which approach would best suit you?

Answered: 53 Skipped: 0

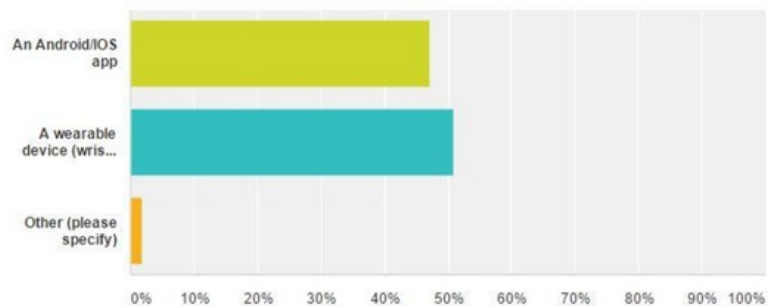


Figure 75; survey results 6.

How much would you be willing to pay for such a device?

Answered: 51 Skipped: 2

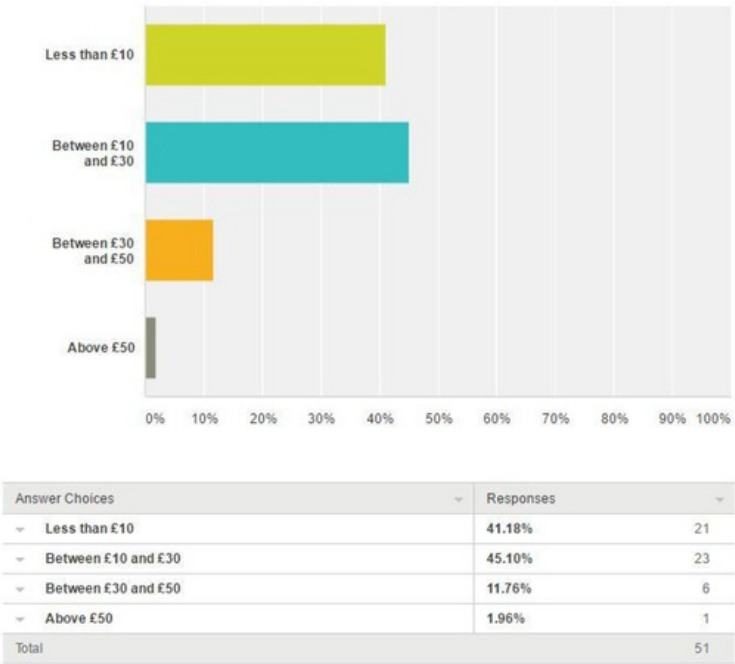


Figure 76; survey results 7.