UNIVERSITY OF GHANA
COLLEGE OF BASIC AND APPLIED SCIENCES
SCHOOL OF ENGINEERING SCIENCES
DEPARTMENT OF BIOMEDICAL ENGINEERING

DESIGN OF A WEARBLE HOLTER FOR EARLY DETECTION OF ARRHYTHMIA

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THIS THESIS IS SUBMITTED TO THE UNIVERSITY OF GHANA, LEGON IN PARTIAL FULFILMENT OF THE AWARD OF MASTER OF PHILOSOPHY IN BIOMEDICAL ENGINEERING

SEPTEMBER, 2024

DECLARATION

I, George Akwasi Amponsem declare that this project work except references to other people's works which have been duly listed was the result of my original work under the supervisions of Dr. Srinivasan S. Balapangu and Dr. Margaret Richardson, and that it has neither in part nor in whole been presented for the award of degree or certificate in University of Ghana.

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ABSTRACT

Cardiovascular diseases (CVDs) remain a cause of death globally, impacting over 17 million people annually. Among the complications associated with CVDs, arrhythmias, which refer to irregular heart rhythms that can manifest as the heart beating too fast, too slow or erratically. These irregularities can lead to life-threatening events such as stroke or sudden cardiac death if not promptly identified and addressed. Early detection of arrhythmia is critical, but current diagnostic tools like Electrocardiograms and Holter monitors are limited in terms of accessibility, short monitoring durations, and lack of real-time data transmission.

While wearable devices offer continuous, real-time monitoring for early detection of arrhythmia, they are often scarce and expensive in low-resource settings like Ghana, where patients frequently make multiple hospital visits for heart monitoring. To address these challenges, this study aims to develop a cost-effective wearable Holter monitor.

The proposed system was developed through three phases: design, software integration and testing. ECG signals are continuously captured using disposable electrodes, processed via a microcontroller, and transmitted onto a mobile application for real-time visualization, alerts and storage.

The wearable Holter monitor was evaluated using an ECG patient, and it demonstrated an average heart rate of 79.8 bpm with a mean deviation of ±0.4 bpm indicating precision at significant levels. Additionally, the device effectively detected both normal and abnormal heart rhythms across a variety of conditions. The lightweight and unobtrusive design improves patient comfort, making it suitable for extended use. By improving accessibility and comfort, this wearable Holter monitor has the potential to significantly enhance arrhythmia detection and patient care, particularly in resource-limited settings.

Keywords: Cardiovascular Diseases (CVDs), Arrhythmia, Holter monitor, Electrocardiogram (ECG).

ACKNOWLEDGEMENT

I find myself overwhelmed with gratitude as I reflect on my postgraduate journey and the incredible support I have received along the way. First and foremost, I owe an immense debt of gratitude to the Almighty for guiding me through this challenging yet rewarding chapter of my life.

I would also like to thank my family and friends, who in diverse ways supported me morally, financially, and spiritually. Special thanks to my boss and colleagues from the Ghana Health Service, who graciously filled in for me during times when I had to attend school. Their support allowed me to balance my professional and academic responsibilities, and for that, I am deeply appreciative.

My heart is full of appreciation for my supervisors, Dr. Srinvasan S. Balapangu and Dr. Magaret Richardson, and Miss Jamila Bayaba. Their mentorship and consistent support were like beacons, guiding me through the academic maze.

I am also grateful to all the faculty members and Senior staff at the department of Biomedical Engineering who in various ways impacted my education and made contributions to my training.

I want to highlight the incredible assistance from Mr. Nicholas Tali from Ashesi University and Gideon Mba from Polymorph. Their expertise and guidance in navigating complex engineering design challenges were invaluable. Without their help, I wouldn't have overcome those hurdles.

Finally, I would love to thank all my loved ones, colleagues, and friends, especially Bernice Okrah and Felix Sarfo for their encouragement, criticism, advice, and support throughout my entire two years of education at the University of Ghana. Your impact has been profound, and I feel immensely blessed to have such wonderful souls in my life.

Thank you, from the bottom of my heart.

DEDICATION

This work is dedicated to the Almighty God for His abundant grace, mercy, and compassion over my life, and the lives of my family, friends, colleagues, lecturers, and loved ones.

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CHAPTER ONE

INTRODUCTION

1.1 Introductions

Over 17 million people die from cardiovascular diseases (CVDs) each year, making them the leading cause of death worldwide(Mensah et al., 2019). In Ghana, the impact of CVDs is worsened by difficulties in obtaining essential medical equipment and skilled healthcare practitioners, especially in the Northern area. It affects 1% to 2% of adults in Ghana, which means that 100,000 to 200,000 people here have cardiovascular diseases that have been diagnosed, and it is believed that many more have undiagnosed CVD. These numbers increase dramatically in individuals over 70 years old, with over 10% being affected(Agyekum et al., 2023; Amoah & Kallen, 2000; James et al., 2018). This is particularly concerning as our population continues to age, presenting a major challenge for modern medicine and health economics.

Arrhythmias have become one of the most important causes for concern among the various cardiovascular disease-related complications. The term "arrhythmia" refers to variations in the heart's rhythm. It describes any deviation from the regular flow of electrical impulses that the heart uses to control how hard it pumps. The symptoms can be quite varied, ranging from weakness and lethargy to dizziness and even fainting. These irregular beats can show up as tachycardia (too fast) or bradycardia (too slow), and their severity level may vary from being a minor issue to a fatal situation. In the case of severe forms of arrhythmias, there is ventricular fibrillation which if not treated immediately leads to the death of a patient.

Atrial fibrillation (AF), another form of arrhythmia, is common and a major risk factor for stroke. It occurs when different atria contract at the same time, causing the muscle to quiver or fibrillate uncontrollably. Nonetheless, atrial fibrillation can lead to other severe complications such as heart failure and stroke, which may be life-threatening but are usually not considered among the leading causes of death (Mensah et al., 2019; Yuyun et al., 2020).

Different cardiovascular diseases have several risk factors such as age, hypertension, diabetes, hyperlipidemia, smoking, obesity, and family history that increase their prevalence rates (Chugh et al., 2014; Kirchhof et al., 2016). These elements may increase the risk of cardiac arrhythmias, which would increase the complexity and seriousness of the corresponding cardiovascular conditions, especially as global populations age and lifestyle changes. According to the findings of a study that was carried out in a tertiary hospital in Accra, Ghana, patients who were being evaluated for cardiac arrhythmia had the presence of potentially fatal arrhythmias in 42.3% of the cases (Doku et al., 2022).

Early detection and treatment of arrhythmias are important, as they can be difficult to diagnose because they often have subtle and intermittent symptoms like mild dizziness or heart palpitations, which patients often attribute to stress and fatigue. Nevertheless, these seemingly harmless signs could be signs of an underlying arrhythmia that needs to be treated. Besides, the ability for early detection enables one to act immediately, slowing down the progress of heart disease while reducing the chances of complications that render one disabled (García-Fernández et al., 2016; Oude Wolcherink et al., 2023; Parish et al., 2018; Stegmann et al., 2022).

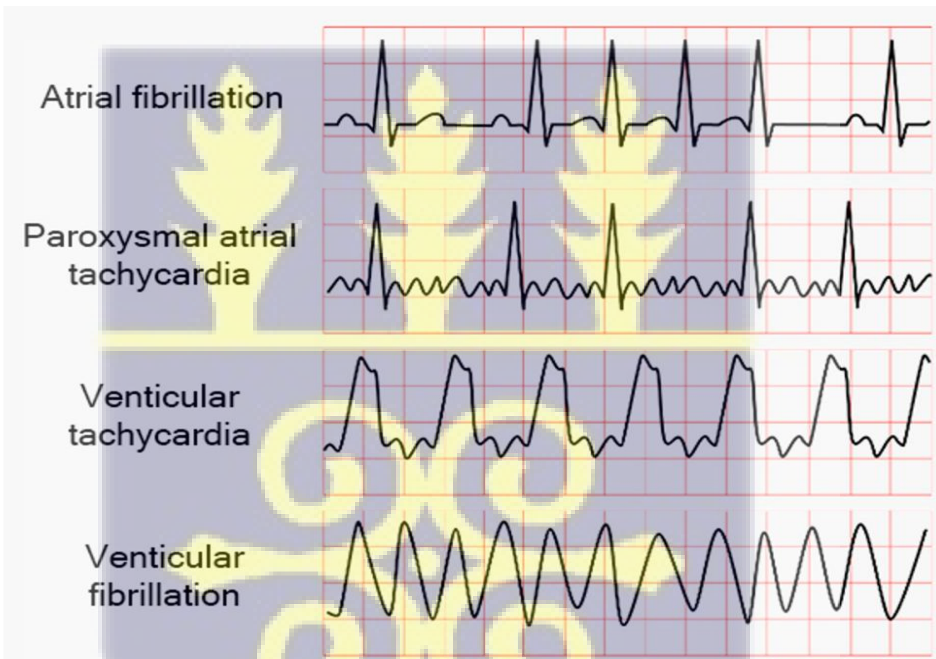


Figure 1. Types of Arrhythmias (Center, 2024).

The diagnosis and understanding of cardiac arrhythmias heavily rely on recording the electrical activity of the heart. Historically, the study of heart rhythm disorders dates back to ancient China and Egypt, with significant advancements in the 19th century when the heart's electrical activity was first discovered (Fye, 1987; Janse & Rosen, 2006; Lüderitz, 2003, 2009). Today, the conventional method for detecting arrhythmias is through a 12-lead electrocardiogram (ECG), which is a thorough test that provides in-depth information about the heart's electrical activity(Solomon et al., 2016). Although this method helps detect arrhythmias during the monitoring period, it may miss arrhythmias that take place outside of the timeframe that is being monitored(Carrington et al., 2022). This limitation becomes especially important when attempting to diagnose infrequent or intermittent arrhythmias, which might not occur at the same time as the short monitoring duration. Again, ECG machines are typically only found in healthcare facilities, and this requires patients to make multiple visits to a hospital or clinic to receive ECG monitoring. This can be inconvenient and time-consuming for the patient, leading to lower compliance and missed opportunities to detect arrhythmias in their early stages.

These shortcomings prompted the development of more sophisticated monitoring technologies like event recorders, Holter monitors, and Implantable Cardioverter Defibrillators (ICD) which provide continuous and longer-term monitoring. They make it easier to find and diagnose arrhythmias as the patient goes about his or her regular daily activities (Kennedy, 2013).

ICDs are small devices that are surgically implanted in the chest, and are capable of detecting arrhythmias and delivering electric shocks to the heart to restore a normal rhythm. ICDs are typically used for patients who are at high risk of sudden cardiac death (Priori et al., 2015; Zimetbaum & Goldman, 2010). Event recorders allow patients to capture their heart activity during specific symptoms. They aid in diagnosing arrhythmias that are triggered by specific activities or events(Lauren, 2021). On the other hand, Holter monitors continuously record the electrical activity of the heart for 24 hours or longer. They are small, portable devices that run on batteries; they are usually worn or attached to the patient using ECG electrodes(Kennedy, 2013). They can collect data from up to 12 different points on the body to record the electrical activity of the heart. The recorded information is then stored on the device using either traditional analog storage or modern digital storage methods. The reading of the heart activities obtained is later taken to a healthcare professional to analyze the recorded information, spot any abnormalities, and determine the seriousness of the condition. This test is painless and doesn't require any invasive procedures, allowing doctors to gather important information and make well-informed decisions about further tests or suitable treatments for the patient's heart health. In recent times, Holter monitors have provided continuous monitoring of the patient's heart activities for 24 hours to a maximum of 72 hours(Forouzanfar & Cafazzo, 2016; Nofitasari et al., 2020).

Wearable Holter monitors are modern development that offer continuous heart monitoring while being more comfortable and discreet than traditional Holter monitors. These wireless devices allow patients to maintain mobility during monitoring and provide privacy, as they are less visible without bulky wires and cables (Adasuriya & Haldar, 2022).

Despite their advantages, the use of Holter monitors in healthcare facilities in Ghana faces challenges primarily due to factors such as inadequate healthcare infrastructure and financial limitations. Even Teaching Hospitals which are the referral points for the other levels of healthcare providers in Ghana have limited availability of this medical equipment. The scarcity of Holter monitors further intensifies the difficulties experienced in hospitals across the various districts and suburbs, particularly affecting the monitoring of cardiac conditions and the provision of treatment to patients.

1.2 Problem Statement

Arrhythmias pose a significant public health challenge, as early detection is crucial for timely clinical intervention and improved patient outcomes. Continuous cardiac monitoring is especially important for identifying intermittent rhythm abnormalities that may not be detected during routine examinations. In Ghana, however, access to effective and continuous cardiac monitoring technologies remains limited.

Holter monitors, which are essential for prolonged arrhythmia detection, are scarce even in teaching hospitals that serve as major referral centers. In addition to their limited availability, most Holter monitoring systems currently in use only support offline data recording, without real-time monitoring or transmission. This limitation restricts prompt clinical decision-making and increases the likelihood that transient or life-threatening arrhythmic events may go undetected during monitoring periods.

Due to the high cost and limited deployment of Holter monitors, many healthcare facilities rely heavily on conventional ECG machines, which are themselves insufficient in number, particularly in district hospitals. This results in long waiting times, repeated appointment rescheduling, and delayed diagnoses, making the detection of intermittent arrhythmias difficult. Together, these challenges highlight a critical gap in cardiac care infrastructure in Ghana and underscore the urgent need for affordable, real-time, and scalable cardiac monitoring solutions.

1.3 Justification

Therefore, there is a pressing need to investigate alternative solutions that can overcome the current scarcity of Holter monitors within Ghana's healthcare system, thereby improving arrhythmia detection and prompt intervention. This will lead to more effective treatment, reducing potential cardiac risks and ultimately improving patient outcomes.

1.4 Research Question

Can the integration of cloud-based storage and real-time mobile app visualization in a Holter monitor system enhance the early detection of arrhythmia and facilitate timely intervention?

1.4 Aim

The aim of this study is to design a low-cost wearable Holter monitoring system for continuous cardiac monitoring, to aid in the early detection of arrhythmias.

1.5 Objectives

The goal of this project is to develop an alternate Holter monitoring system that supports cloud-based storage and real-time mobile app visualization features. This design will help improve early detection of arrhythmias and facilitate timely intervention, ultimately enhancing patient results in the management of cardiac rhythm disorders. The project is divided into three main phases, each of which focuses on achieving particular objectives.

- Objective 1: To design and build a wearable Holter monitoring system capable of continuous ECG signal acquisition.
- Objective 2: To design and develop a mobile app that wirelessly streams data from the Holter monitor, and graphically displays it.
- Objective 3: To evaluate the performance of the proposed system using an ECG simulator.

By achieving these goals, this project seeks to advance arrhythmia monitoring technology by offering an effective and trustworthy tool for arrhythmia diagnosis. The Holter monitor and smartphone app have the potential to improve early detection and diagnosis of arrhythmia, resulting in more efficient treatment plans and better patient outcomes.

CHAPTER TWO

LITERATURE REVIEW

2.1 Bioelectrical Signals

Bioelectrical signals are associated with the ionic processes that occur due to electrochemical activity in specific excitable cells, such as those found in the brain, heart, and muscles. These cells possess the capacity to create action potentials by altering ion concentrations across their membrane, resulting in the flow of current in the surrounding tissue. This electrical activity can be measured intracellular or extracellular using electrodes (Enderle et al., 2011).

To record a bioelectrical signal, it is necessary to attach at least two electrodes in order to detect potential changes. Various electrode configurations are frequently employed to provide a spatial representation of the biopotential phenomena. However, it is generally not possible to get specific information about cells directly because the signals are recorded from a distance, with different tissues in the way.

An electrocardiogram (ECG) is a significant instance of a bioelectrical signal that captures the heart's electrical activity, recorded on the chest and limbs, see Figure 2. The distinct sequence of PQRST complexes in the electrocardiogram (ECG) signal correlates to the repetitive stages of activation and recovery of the heart (HeartRhythDoc., n.d.).

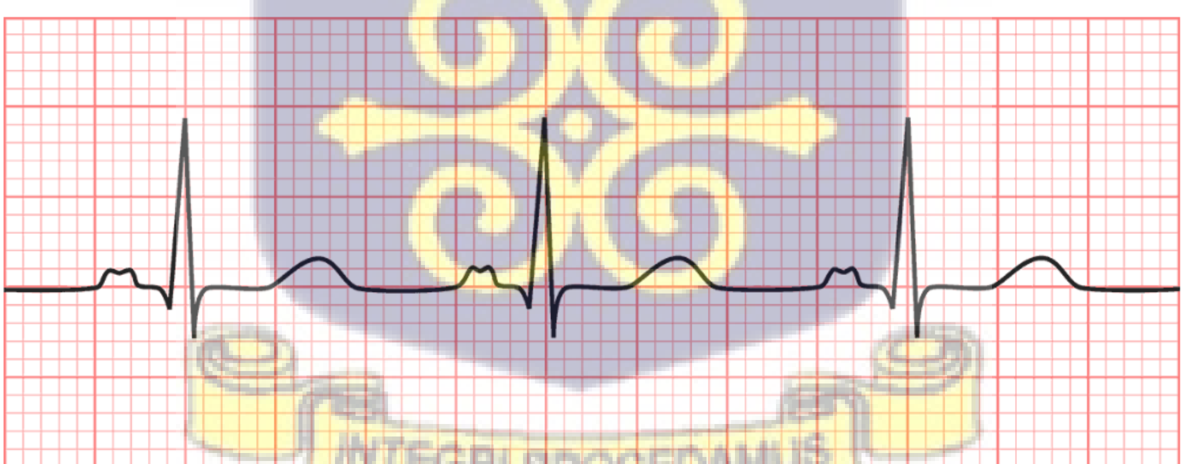


Figure 2. An example of bioelectrical signal - an electrocardiogram (ECG), recorded (HeartRhythDoc., n.d.).

2.2 Electrical Phenomena of the Heart

The heart's electrical activity is dependent on variations in the levels of potassium, sodium, calcium, and chloride ions both inside and outside the cells of the heart muscle(Wei et al., 2023). During the resting state of a cell, the cell membrane exhibits polarization, characterized by a potential difference of around -80 mV between the internal and external environments. This polarization is caused by an unequal distribution of ions across the membrane(Grant, 2009).

When the cell receives an electrical stimulation that exceeds a certain threshold, it quickly undergoes depolarization as sodium ions enter, changing its charge to positive. This marks the zero phase of the action potential (AP). The subsequent repolarization phases (1-3) encompass various movements of potassium, calcium, and chloride ions, finally reinstating the negative resting potential(Lin et al., 2023).

The transfer of these action potentials stimulates the myocardium at the same time. The impulse originates in the sinus node pacemaker cells and proceeds across the conduction system, passing through the AV node, His bundles, bundle branches, Purkinje fibers, and atria before activating the contraction of the ventricle.

Electrodes placed on the skin can detect an electrocardiogram (ECG) signal when certain parts of the heart are repeatedly stimulated(Fu et al., 2020). The PQRST waves reflect the depolarization and repolarization of the atria and ventricles. The bioelectrical signal plays a crucial role in the noninvasive detection of various cardiac conditions.

2.3 Heart Rate

Heart rate, sometimes referred to as pulse rate, is a measurement of how often your heart pumps blood around your body in one minute. The average healthy adult's resting heart rate falls between 60 and 100 beats per minute(Kang et al., 2017). Heart rate, however, might differ based on several variables, including age, degree of exercise, physical condition, and others. An underlying health condition may be indicated by a resting heart rate that is outside of the normal range; this condition should be assessed by a medical practitioner.

There are several methods of monitoring heart rate, including electrocardiography (ECG), which uses electrodes attached to the skin to measure the electrical activity of the heart. Blood pressure can also be monitored using a blood pressure cuff to measure the changes in pressure of the blood flowing through the arteries. A cuff-less blood pressure monitor that measures real-time blood pressure using a microphone or photoplethysmography (PPG) sensor can also be used to measure heart rate. PPG uses a light source and a photo-detector at the skin's surface to measure the volumetric variations in blood. It is a non-invasive technology(Ghamari, 2018; Pour Ebrahim et al., 2019). Heart rate is then extracted from the body signals either by frequency domain analysis, time domain analysis, machine learning algorithms, or hybrid approaches by combining any of the three(Bansal et al., 2009; Bhoi et al., 2018).

2.2 Electrodes

Electrodes are devices utilized for measuring the electrical activity of the body, and they can be broadly classified into two types: surface electrodes and invasive electrodes. As their name suggests, surface electrodes are applied to the skin, whereas invasive electrodes are inserted into the body. Although there is a small chance of infection, invasive electrodes offer extremely accurate measurements. Surface electrodes, on the other hand, are much easier to use and do not put the patient at any real risk. Surface electrodes can be divided into the following subcategories:

- Disposable Adhesive Electrodes: These electrodes consist of a conductive adhesive surface that adheres directly to the skin. They typically contain a conductive gel or solid gel layer and a metal chloride embedded in the center.
- Suction Electrodes: Suction electrodes employ small cups filled with conductive gel that are pressed against the skin. The suction mechanism helps maintain consistent and reliable contact between the electrode and the skin surface.

These electrodes are commonly used in various medical procedures and diagnostics, such as electrocardiography (ECG), electromyography (EMG), and other monitoring applications. These surface electrodes offer convenience, ease of application, and accurate signal acquisition without the need for invasive techniques.

2.3 ECG and Heart Rate Monitors (elaborate)

The electrocardiogram (ECG) is a vital diagnostic tool for measuring and monitoring the rhythm of the heart's electrical activity. The electrical activity is initiated by a group of cells called the sinoatrial (SA) node, which is located in the right auricle (see Figure 3) (Standford Children's Health., n.d.). This signal then travels through the right and left auricle causing the auricles to contract and pump blood into the ventricles. This electrical impulse is captured on the ECG as a P wave. The impulse moves through the auricles and ventricles through a group of cells called the atrioventricular (AV) node. Here the impulse is momentarily stalled allowing the blood to fill the ventricles. This is recorded as a flat line between the end of the P wave and the onset of the Q wave in an ECG signal. From the AV node, travel via a pathway known as the bundle of His that branch into the left and right sides of the heart, causing the ventricles to contract thereby pumping blood to the rest of the body. This process is represented by the QRS wave on the ECG signal. A heartbeat can be detected in each ventricular contraction. Just after the QRS wave, the ventricles return to their normal electrical state, represented by the T wave on the ECG signal as shown in Figure 4(Dilshad, 2019). Muscle contraction is then paused to allow the heart to receive fresh blood(Joakim et al., 2007; Taccardi et al., 1997).

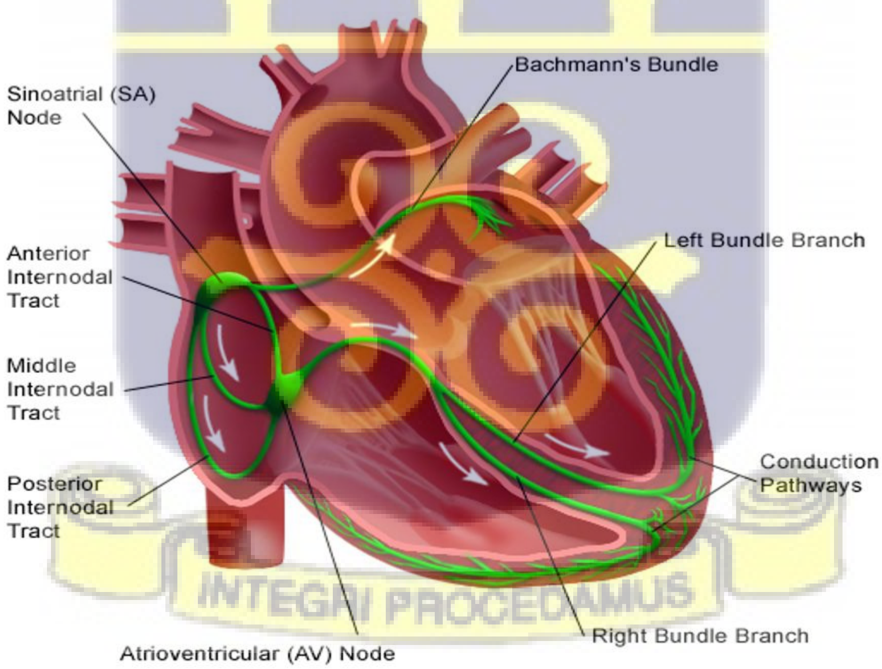


Figure 3. A diagram of the electrical system of the heart showing the four main components of the electrical system: the sinoatrial (SA) node, the atrioventricular (AV) node, the His-Purkinje system, and the conduction pathways (Standford Children's Health., n.d.).

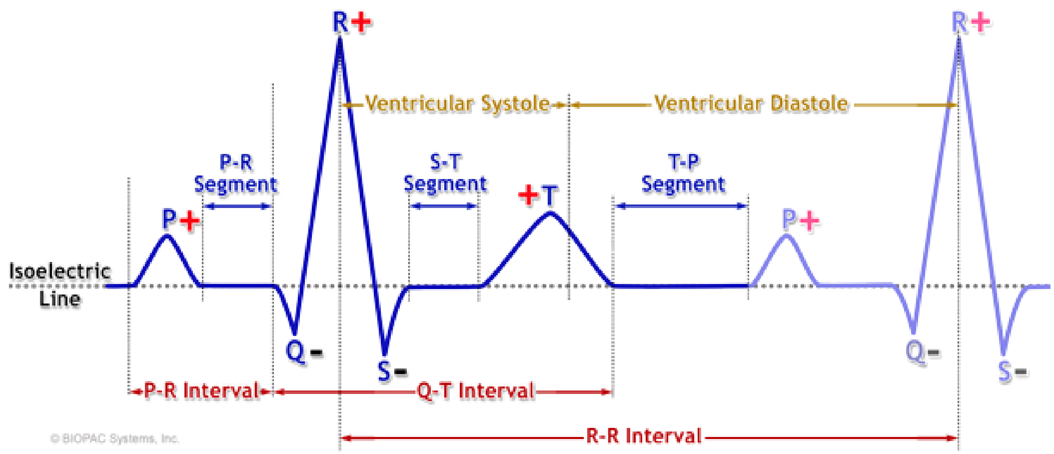


Figure 4. a diagram of a typical ECG (electrocardiogram) showing the different sections. P wave (represents the depolarization of the atria, the upper chambers of the heart), QRS complex (represents the depolarization of the ventricles, the lower chambers of the heart) and T wave (which represents the repolarization of the ventricles) (Dilshad, 2019).

Table 1 Summary of the Electrical system of the heart

ECG SIGNAL	SOURCE OF ECG SIGNAL
P Wave	Atrial Depolarization
QRS Complex	Atrium Repolarization and Ventricular Depolarization
T Wave	Ventricular Repolarization

Table 2. The Average duration of different components of an electrocardiogram (ECG) signal

ECG SIGNAL	DURATION
P Wave	0.08 – 0.10 secs
PR interval	0.12 – 0.20 secs
QRS Complex	0.06 – 0.10 secs
QT interval	< 0.44 secs

The ECG machine measures the heart signals by electrodes affixed to the body's surface. The obtained signal shows a comparison of the voltages recorded by two electrodes that are positioned at various locations on the body. The term "leads" can also be used to refer to these pairs. Standard

ECG configuration applies 1, 3, 5, or 12 leads in acquiring electrical signals of the heart. Figure 5 displays the standard 12-lead electrode placement (Sherwood & Christopher, 2016).

Although the conventional way to detect arrhythmia is through a 12-lead ECG, which is a more detailed test that provides in-depth information regarding the heart's electrical activity, it is typically available only in healthcare facilities and cannot be used in a field setting due to its size and cost(Aimcardio Team, n.d.).

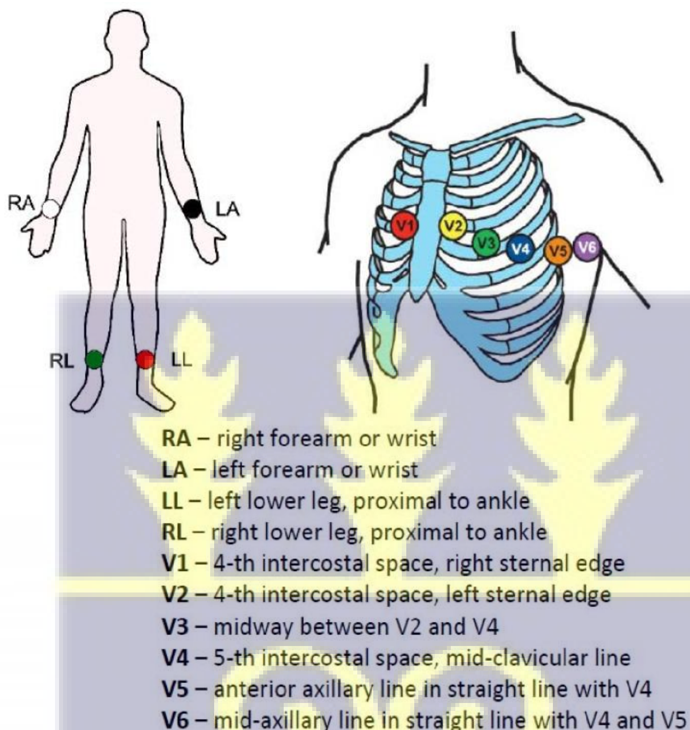


Figure 5. 12 Lead ECG Electrode Placement (Sherwood & Christopher, 2016).

In recent times, research findings and developments have discovered a more comfortable, mobile, and effective way of continuously monitoring heart activities using Holter monitors.

Holter monitoring devices run on batteries and can continuously track different cardiovascular system activities for more than 24 hours (Tanaka et al., n.d.). It has electrodes that are fastened to the patient's chest while the equipment records the ECG and can function as an ECG recorder that is strapped to a person and carried about at all times. The electrical impulses from the ECG are picked up by the electrodes and amplified so that a doctor can read and diagnose them(Forouzanfar & Cafazzo, 2016). The device is capable of transmitting the recorded data to the user or the healthcare officer once the recording time has expired or there is an abnormal reading. The integration of digital medical data and electronic health records into daily life through wearables and smartphone applications has generated enormous amounts of data which has transformed healthcare in many ways. The availability of these innovations and their uses is growing, and they show promise for changing medicine(Han, 2020).

Furthermore, recent research demonstrates that Lead II ECGs obtained using Holter devices are comparable to those obtained using conventional 12-Lead ECGs in terms of known ECG traits, such as R peak location and P-wave amplitude(Dörr et al., 2019; Isakadze & Martin, 2020; Pierleoni et al., 2019).

2.4 Current Standards of Care for Arrhythmia Detection

Short-term ECG recording and ambulatory ECG monitoring are the current standard methods for detecting arrhythmia (Rosenberg et al., 2013). The most common first-line method for detecting arrhythmia is by the use of a 12-lead ECG which necessitates the patient to be at rest. This is usually used in a clinical setting. It gives a snapshot of the heart rhythm and electrical activity at the point of care. A standard 10-second recording of the 12-lead ECG has an average sensitivity of 75%-95% in detecting arrhythmias such as atrial fibrillation but has a lower sensitivity for detecting other arrhythmias (January et al., 2014). This is due to the short recording time, which may fail to capture asymptomatic arrhythmia episodes due to their episodic nature (Turakhia et al., 2015).

Holter monitoring is used to detect sporadic events, in which patients continuously wear an ECG recording device for 24-48 hours. Nevertheless, it's possible that even 48 hours won't be enough to catch irregular heartbeats. The primary drawbacks are the brief monitoring period, which might not pick up very rare arrhythmic episodes, and the inability to offer real-time monitoring and alerts (Rosenberg et al., 2013; Steinhubl et al., 2018).

Implantable loop recorders can be used for long-term monitoring for up to three years, but they require minor surgical implantation and come with associated costs and risks, such as infection. Data can only be retrieved after patient activation or remote transmission at predetermined intervals, so they do not allow for real-time monitoring (Reiffel et al., 2005). To improve early detection of arrhythmias, there is a need for intermediate-duration, non-invasive monitoring.

2.5 Research on existing wearable ECG/ Holter Monitors

The development of the Holter monitor by Norman Holter in 1961 marked the beginning of a revolution in the field of mobile cardiac monitoring. His initial device recorded ECG during ambulatory monitoring on analog tape (see Figure 6) (Holter, 1961). These early Holter monitors, however, were extremely inconvenient because of their dependency on tape recorders, which were restricted to only 24 hours of discontinuous recording and lacked signal processing qualities. Subsequent Holter research concentrated on reducing the size of the devices and increasing their level of digitalization. Although cassettes and smaller recorders were reported by Kennedy, the usage of chest cables and electrodes was still a nuisance(Kennedy, 1992). According to Mittal et al.'s review in 2011, solid-state flash memory started to take the place of tape by the 1990s (Mittal et al., 2011).

Figure 6. Holter Electrocardiocorder unit by Norman Holter in 1961 with a tape recorder, carrying case, and coupling electrodes (Holter, 1961).

In clinical investigations evaluating classic Holters, researchers discovered that while the tests were beneficial for the diagnosis of short-term arrhythmias, they also had substantial limits. These limitations were caused by the short monitoring periods of only one to three days. The problems with such brief monitoring periods are highlighted by the findings of Joshi et al. and Kulach et al., who observed that 30-day event monitors recorded arrhythmias in roughly a third of patients in whom a preceding 24-hour Holter revealed no occurrences (Joshi et al., 2005; Kułach et al., 2020).

Further studies were conducted to improve traditional Holters, including new lead configurations (Stracina et al., 2022), integrated buffer circuits in electrodes (Beach et al., 2021), and ultra-low-power analog ASIC chip designs (Yazicioglu et al., 2011) to provide greater resolution of cardiac activity, reduce signal artifacts during motion, and prolong battery life. However, the underlying limitations regarding wearability, duration, and patient compliance continued to be significant issues highlighted by several studies.

To circumvent the inherent limits of traditional Holter technology, more recent research has concentrated on designing wearable ECG monitoring systems. These include novel adhesive patch devices with miniaturized electronics that can constantly monitor heart activity for weeks rather than days (Kwon et al., 2021; Walsh et al., 2014).

2.6 Gaps in Current Technology and Knowledge

In the context of early research on wearable ECGs, several studies identified issues with data quality and signal artifacts. In 2011, Ramli et al. designed a low-cost heart beat monitoring device that used reflectance mode photoplethysmography to extract pulse signal from the finger, but the accuracy and efficiency remained an issue as the circumference of the user's finger were proven to affect accuracy (Ramli et al., 2011). Again, Poh et al. encountered accuracy issues with the initial version of their wearable heartbeat monitor prototype (Poh et al., 2011). A commercial heart rate monitor was also validated by Dolezal et al. against an ECG. Though the monitor was able to measure the heart rate, it was unable to assess cardiac rhythm (Dolezal et al., 2014).

In more recent wearable research, electrocardiogram (ECG) capabilities have been included for ambulatory arrhythmia screening. Mahdy et al. in 2018 created a simple Holter system based on an Android mobile application for the detection of arrhythmias in real-time(Mahdy et al., 2018). The recorded signals are transmitted through Bluetooth to the mobile device. However, in situations where the user lacks access to their phone, data loss occurs, posing a challenge as the monitoring period is affected. Demircioglu et al. designed an inexpensive ECG patch capable of basic threshold-based alerts but relying only on heart rate rather than complex waveform analysis. Even though heart rate plays an important role in detecting arrhythmia, it is difficult to ascertain the type of arrhythmia as the rapid heartbeat may be due to the user undergoing vigorous exercise(Demircioglu et al., 2022).

In 2020, D. Nofitasari et al. (Nofitasari et al., 2020) developed a portable and low-cost Holter system to record the ECG signal in twenty-four hours. The ECG signal is acquired from the body using the LEAD II standard measurement. An SD Card Memory was employed to capture the raw data from the ECG signal and save it for later data processing. A Bluetooth transmitter was also included in the design to send the data to a computer. Even though the accuracy of this device allows for real-time ECG signal monitoring, there is a great chance of missing arrhythmia since the data is recorded for future analysis.

Additionally, the availability and cost of wearable Holter monitors may be restricted in countries with lower incomes or in rural areas, where access to advanced medical technology and resources is limited. The price of these devices can have a big effect on how widely they are used and how many people buy them. Hughes et al. emphasized the limited availability of Holter monitors in developing nations as a result of their high price and lack of infrastructure. The lack of availability makes it difficult for early detection and prompt action, which presents a problem for healthcare in these areas (Hughes et al., 2023).

In light of the issues with the previous system, the objective of this study is to design a wearable Holter for early detection of arrhythmia. The key element of this design is the ability to wirelessly transmit the recorded ECG data to the cloud or to a mobile phone through Wi-Fi, where it can be accessed by the user using a mobile application. The device will also incorporate offline data storage where the generated data will be stored on an SD card. The device will collect ECG signals from the body using the Lead II standard measurement. To extend the monitoring period, it will once more involve the usage of a low-power microcontroller. The suggested model will contain an alarm system to alert the user anytime the heart signals are aberrant.

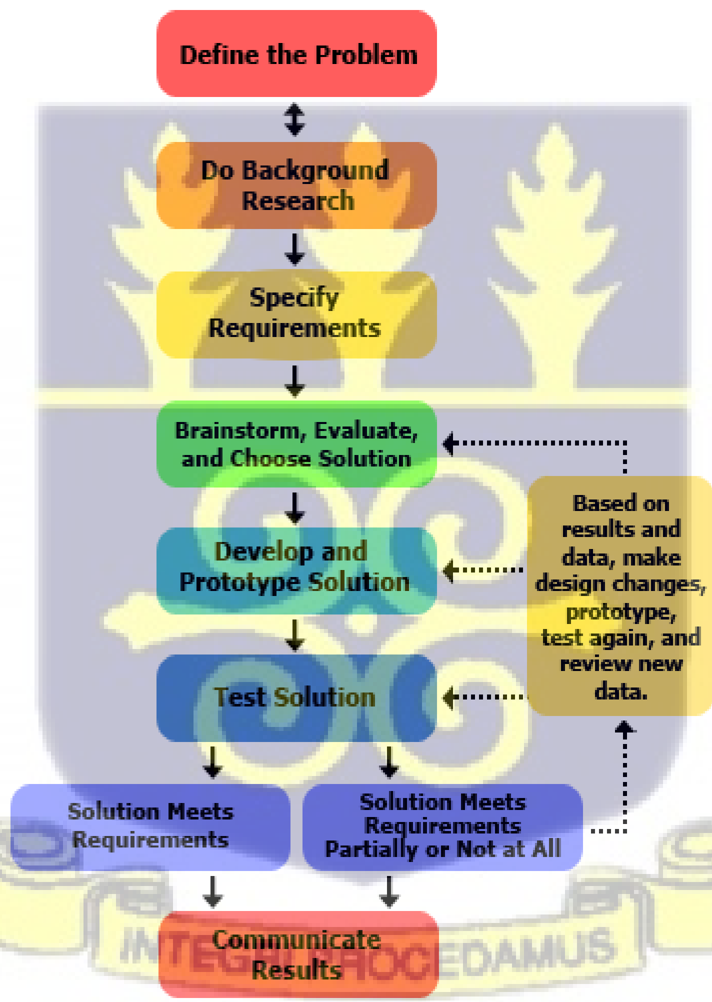
CHAPTER THREE

MATERIALS AND METHODS

3.1 PHASE 1: HARDWARE DEVELOPMENT

3.1.1 Design and Development Approach

The approach used was the Engineering Design Process developed by the Accreditation Board for Engineering and Technology (ABET) to address the need identified and to strictly follow the design process(Buddies, n.d.). The designed process is summarized below.



The first step of the project was to recognize a problem which was accomplished during a lecture at the department. Following the class discussion, the need for the project was determined through consultations and interviews with cardiologists and other medical professionals. We conducted a literature review to look into the current solutions on the market, and from the results, we came up with our functional requirements and specifications. These specifications and requirements acted as a roadmap for the design process, ensuring that the ultimate solution adequately addressed the identified problem. The specifications were created, along with an objective tree highlighting all the functional requirements. To test the design outputs, functional diagrams, schematics, and a prototype were created.

3.1.2 Functional Analysis

To achieve the goals of early arrhythmia identification, our Holter monitor uses a grey box model, which incorporates the signal processing and transmission steps. The model produces medical information on arrhythmias based on raw physiological signals obtained from a sensor. This is depicted by the grey box model of the design in Figure 8.

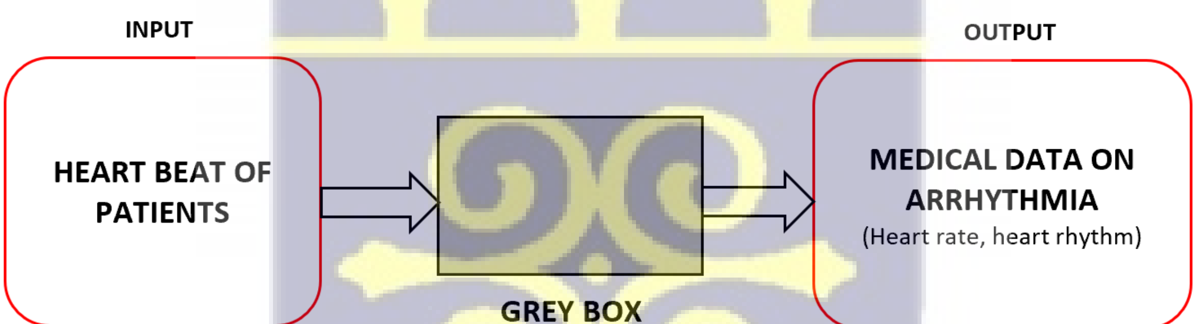


Figure 8. The Grey Box Model of the Holter Monitor (Researchers own construct, 2023)

The functional structure provides a thorough and diagrammatic depiction of the functional analysis (see Figure 9). It also displays the transparent version of the grey box model, overall functions, sub-functions, and input and output characteristics of the Holter monitor.

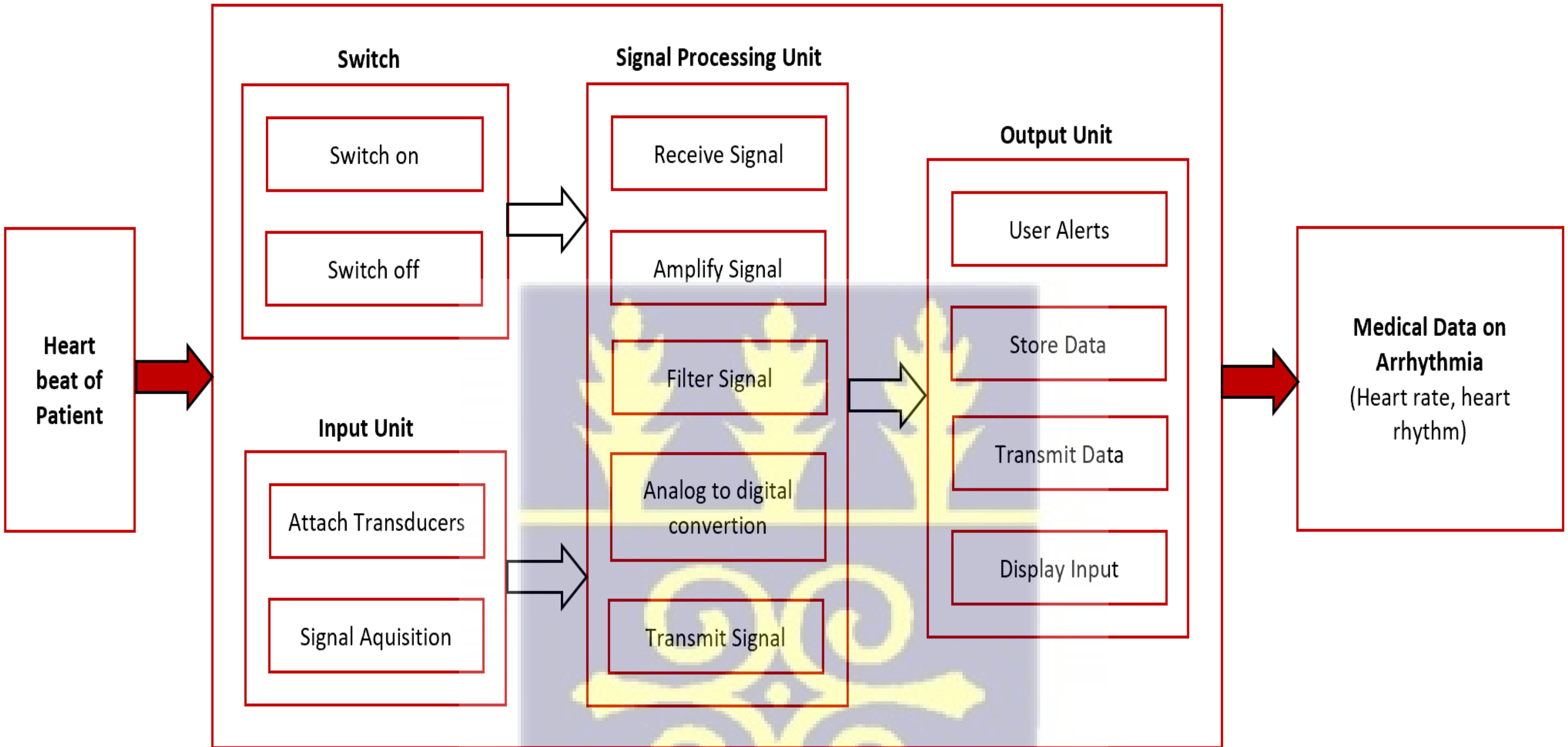


Figure 9. The diagram shows a detailed functional structure of the Holter Monitor. The signal processing unit acquires the heart signals from a transducer attached to the patient's skin where it is filtered and amplified. The amplified signal is converted from analog to digital and then displayed, stored and transmitted (Researchers own construct, 2023).

3.1.3 Design Requirements

The goal of this objective was to design and create a Holter monitoring system that surpasses the limitations of current Holter monitors. That is facilitating the secure storage, access, and retrieval of patient data by medical professionals through the integration of cloud-based storage and real-time monitoring via a mobile app. This allows doctors to quickly and easily access the data from any location, giving them the ability to provide immediate diagnoses and interventions. Based on the findings from the literature review, the requirement of the Holter monitor has been grouped into safety, ergonomics, convenience, and functionality.

Safety:

- Non-Invasive Design: The Holter monitor should be non-invasive to avoid any discomfort or irritation during the extended monitoring period.
- Electrode Contact: The monitor should ensure secure and dependable electrode contact with the patient's skin, ensuring accurate signal acquisition for precise cardiac data analysis throughout the monitoring duration.
- Electrical Safety: To ensure the patient's safety while using the device, the device must comply with applicable electrical safety standards, including elements such as insulation, grounding, and protection against electrical hazards.

Cost Effectiveness:

- The device should be designed with affordability in mind using low-cost materials.
- The overall cost of maintaining the Holter monitor system should be kept at a reasonable level.

Performance:

- The device should be easy to operate and set up by both patients and healthcare professionals.
- The system should be easy to maintain.
- The monitor should support wireless connectivity to seamlessly transfer data to external devices like smartphones or computers without cables.
- The system should be able to precisely record the patient's heart rate and rhythm.

Convenience:

- Data Storage Capacity: The device should have either enough built-in memory or removable storage (such as an SD card) to hold the recorded ECG data for the desired monitoring duration without requiring frequent data transfer or deletion for later analysis by a doctor or other healthcare provider.
- The monitor should be designed to be comfortable, discrete, and unobtrusive, allowing patients to wear it under clothing without drawing unwanted attention or discomfort.
- The Holter monitor must be both compact and lightweight so that it places as little strain as possible on the patient and makes it possible for them to wear it comfortably for the entire duration of the monitoring process while going about their daily activities.
- The monitor should have adjustable straps or fasteners so that it can fit a variety of body sizes while still providing a secure fit.
- Battery Life: To guarantee continuous monitoring, the Holter monitor should have a long-lasting battery that can power the device for the duration of the monitoring without needing to be recharged or replaced regularly.

By meeting these requirements, the Holter monitor aims to provide a seamless monitoring experience for patients while also ensuring the accuracy and reliability of recorded data. In turn, this makes it possible for medical professionals to diagnose patients correctly, analyze cardiac events thoroughly, and recommend the best course of treatment.

Below is an objective tree highlighting all the system requirements.

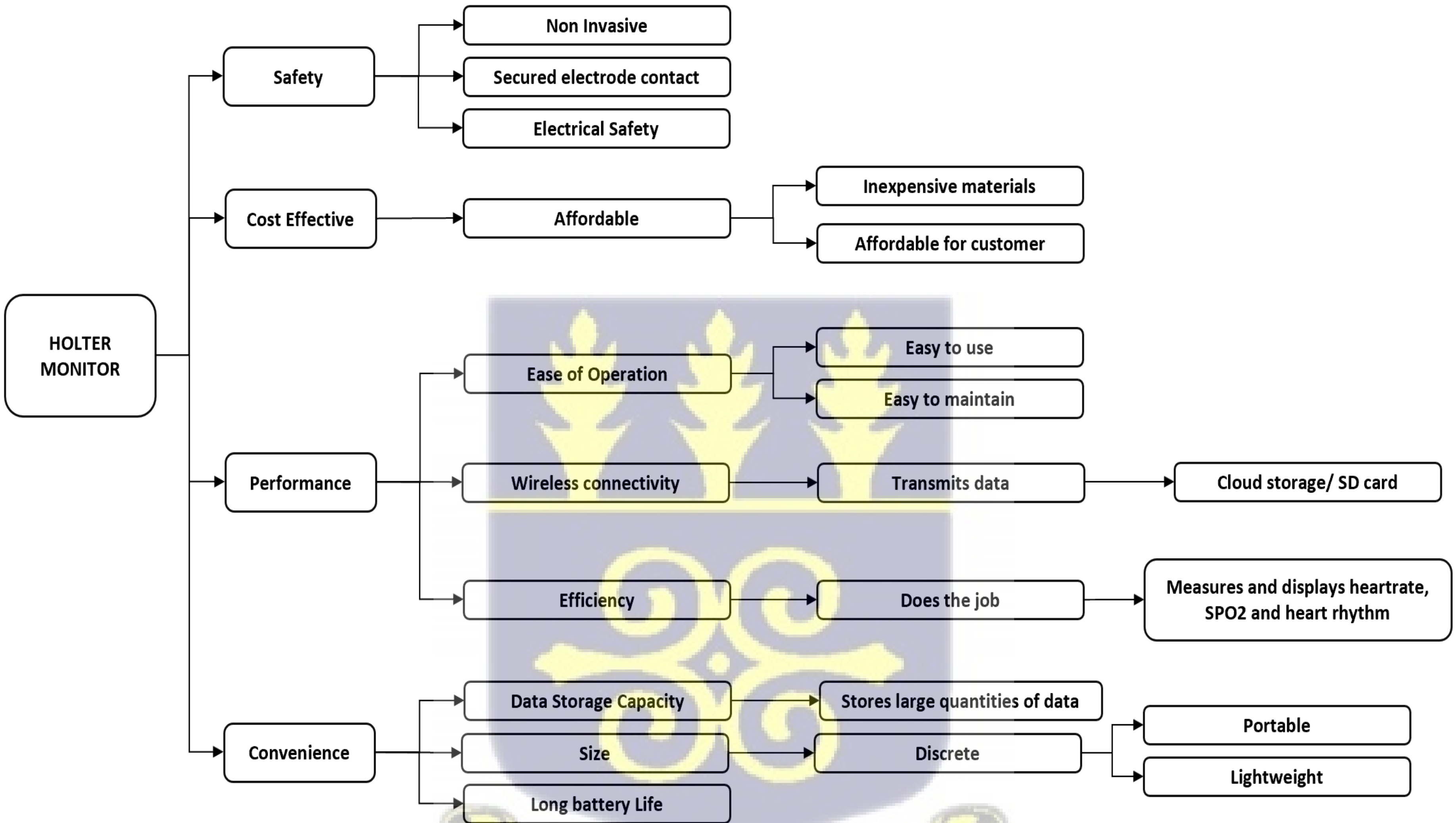


Figure 10. Hardware System Requirements (Researchers own construct, 2023).

3.1.4 Product Specification

A specification checklist presented in Table 3 was developed before the development of a detailed specification (refer to Appendix A, Table 10) that conforms to project objectives and design constraints. However, it must be pointed out that engineering specifications (in Table 3) indicated as demand will be given high priority during design.

Table 3. Hardware Specifications

System Requirement	Demand/ Wish	Engineering Parameter(s)
Detect heart rhythm	Demand	Type of sensor Number of leads
Non-invasive design	Demand	Attachment mechanism Electrode material
Secure and dependable electrode contact	Demand	Electrode material Adhesive strength
Electrical Safety	Demand	Energy Source Insulation Shape and Sharp edges
Discrete	Wish	Size Weight
Affordable	Wish	Material cost Manufacturing cost Component Selection
Comfortable	Wish	Material Size Weight Number of electrodes
Self-Operational	Demand	Automation
Easy to Maintain	Wish	Accessibility of components Software design
Supports Wireless Connectivity	Demand	Wireless communication protocol Data transmission rate
Long Battery Life	Wish	Battery capacity
Data Storage Capacity	Wish	Storage technology Memory
Easy to Operate	Wish	Automation User interface Steps in operation

3.1.5 House of Quality

The House of Quality is a chart that offers a broader perspective of the interactions between engineering parameters and customer needs and highlights the various impacts on achieving the goals of the Holter monitor design. It guarantees that a product is made with the demands of the consumer in mind. Correlation values of 1, 4, and 9 were employed. Requirements that are considered to be highly correlated (9) show that modifications to these engineering features have a significant impact on achieving the related customer requirements. Requirements with a moderate correlation (4) imply a moderate influence on meeting customer requirements based on the specified engineering characteristics. Finally, correlation values of 1 (low correlation) indicate that the chosen engineering qualities may have less influence on achieving these specific customer needs than those with greater correlations (9 and 4). Appendix B. Table 11 displays the house of quality as a table, which offers a more comprehensive understanding of customer needs. An important rating was calculated and included in the specification table (see Appendix A, Table 10) using the data from Table 11.

3.1.6 Conceptualization Using Morphology Chart

In this phase, concepts representing several approaches to the highlighted issue were developed (see Table 4). Concepts were generated through brainstorming and a systematic method that included coming up with alternatives for each function of the proposed solution. The working principles of each function and sub-function were taken into account to determine as many ways as possible to meet the same functional requirement.

Table 4. A table showing several options for subsystems of the Holter Monitor

Features	Concept A	Concept B	Concept C	Concept D
Type of Electrode	Gold	Ag/AgCl	Graphite	Gold
Number of Electrodes	Twelve-lead configuration	Three-lead configuration	Single-lead configuration	Five-lead configuration
Power Source	Rechargeable Batteries	Disposable Batteries	Solar	Disposable Batteries
Fabric	Spandex	Cotton	Cotton	Nylon
Microcontroller	Raspberry Pi	ESP32	Raspberry Pi	Arduino Uno
Casing	Polylactic Acid	Polylactic Acid	Fabric	Plastic
Alert	Audio + Visual	Audio + Visual	Audio	Visual
Storage	Solid State Drive	Mini SD Card	Micro SD Card	Cloud Storage
Display Unit	Smartphone App	Smartphone App and LCD display	OLED display	Smartphone App and OLED display
Switch	Touch	Push Button	Touch	Push Button

3.1.7 Concept Selection Using Decision Matrices

The following "key" was applied to every decision matrix used to choose concepts:

KEY

0: Not Satisfied

2: Fairly Satisfied

4: Satisfied

5: Highly Satisfied

Concept selection is a crucial step when designing a Holter monitor that can accommodate the various demands of both patients and medical professionals. Each design choice was judged and evaluated quantitatively using the decision matrix as a basis. Desirable characteristics from the specifications and literature review were graded and given weights that indicated their significance for each subfunction. The possibilities for the identified subfunctions of the problem were then analyzed using the aforementioned key to determine which solution is most appropriate for implementation. The recommended options were analyzed using technical information from literature, market analysis, and specifications. For each choice, a score is given based on how well it satisfies a desirable attribute. Once all design options had been evaluated following the criteria, the sum of their weighted scores was calculated. The option with the highest score indicates the most suitable choice based on the predetermined criteria. The various identified options were evaluated to choose a final concept (refer to Appendix C).

3.1.8 Materials Selection

Following the selection of the preferred system concept, materials and components were selected to implement the chosen design. Component selection was guided by the design requirements and product specifications derived earlier in the methodology.

Table 5. Selected materials, specifications, and functional roles

Component	Specification	Function
ECG Sensor Module	AD8232	Acquisition, amplification, and filtering of ECG signals
Microcontroller	ESP32	ECG processing and wireless data transmission
ECG Electrodes	Disposable Ag/AgCl	Skin–electrode interface
Display	SSD1306 OLED	Real-time heart rate display
Storage Module	Micro SD Card	Local storage of ECG data
Power Source	9V Alkaline Battery (600 mAh)	System power supply
Voltage Regulators	MC78M05ABDT, REG1117	Voltage regulation
Enclosure	Laser-cut acrylic	Protection of electronic components
Wearable Material	Cotton fabric	Comfort and stability during use

3.1.9 Hardware Design

This hardware design describes the systematic process followed in implementing the electronic and wearable subsystems of the proposed wearable Holter monitoring system. The hardware architecture was developed to support continuous ECG signal acquisition, low power operation, wireless communication and user comfort, in line with the defined requirements and product specifications.

The hardware design was divided into 3 major modules: the electronic module, wearable module and electrodes with each addressing specific functional and ergonomic requirement.

3.1.9.1 Electronic Module Design

The electronic module enclosure houses the electronic circuit board, which is made up primarily of rechargeable batteries for power, an AD8232 integrated signals capable of amplifying and filtering bio-signals, an SD card for storing collected signals when the device is not streaming, and a microprocessor (ESP32) to collect, process, and transmit desired ECG signals from the AD8232. The Central Processing Unit (CPU) processes the ECG signals as they are being gathered in real-time and stores them in a buffer before sending them over radio communication to the smartphone or online storage. Included in the design is an SD card for storing the user's ECG signals and heart rate for future analysis. The architecture of the circuit board is shown in Figure 11.

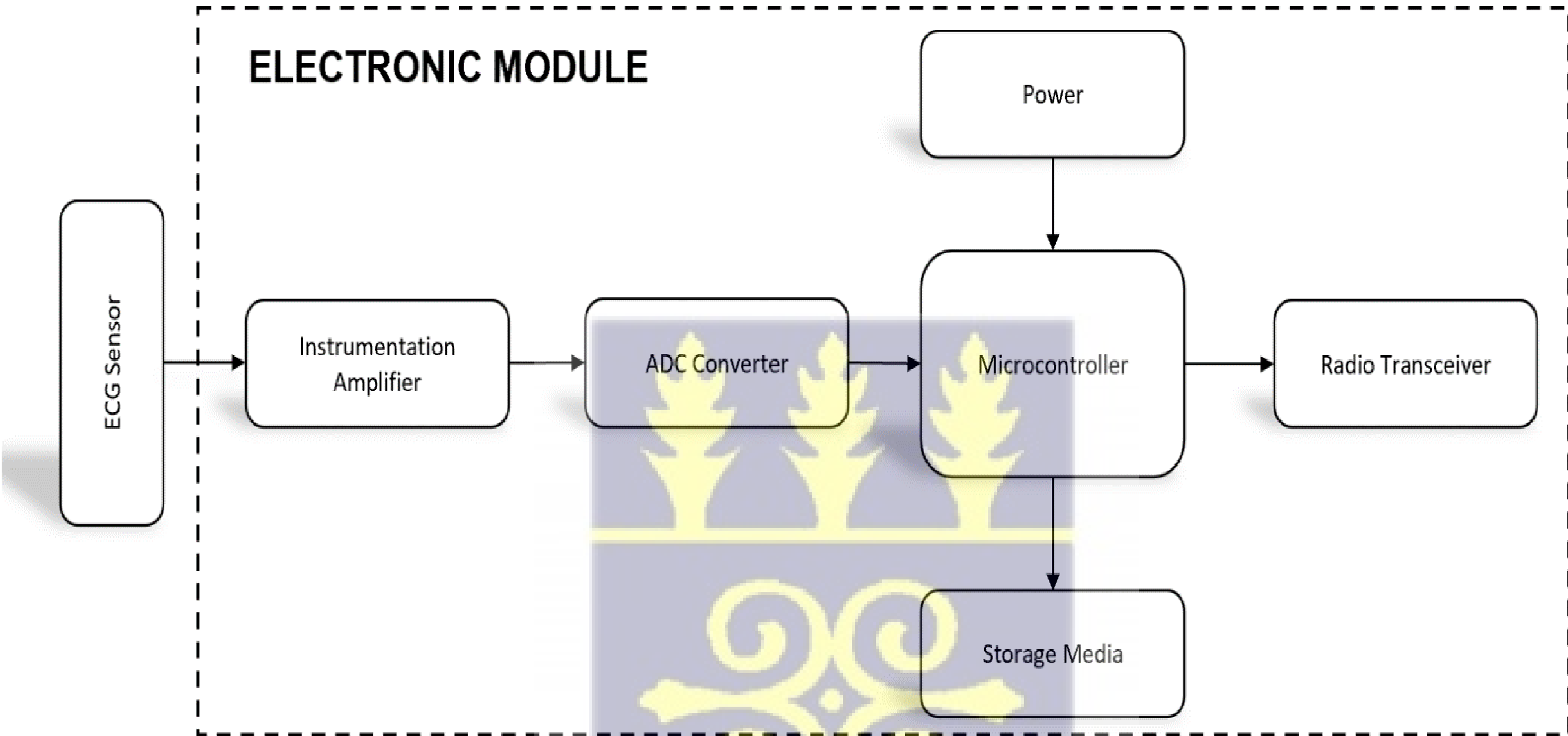


Figure 11. Hardware Architecture of the Electronic Circuit Board (Researchers own construct 2023)

Power

The Holter monitor is powered by a 9V alkaline battery with a capacity of 600 mAh as its power source. We used MC78M05ABDT and REG1117 voltage regulators in our design to ensure that the power supply to the various components is stable and regulated(Management & For, 2004; Mc, 2013). These regulators are responsible for providing a consistent 5V and 3.3V output from the alkaline batteries, respectively.

The MC78M05ABDT is specifically employed in the design to deliver a steady 5V voltage output to the buzzer and display for optimal performance. Similarly, the REG1117 regulator ensures a reliable 3.3V supply, essential for powering sensitive modules like the ESP32 microcontroller and other peripherals.

The Holter monitor can efficiently control power distribution by adding these voltage regulators, ensuring that the components receive the precise voltage they require for functioning. Using an alkaline battery as the power source eliminates the need to charge the gadget, potentially minimizing downtime. This can be useful in instances when continuous monitoring is critical for patient health and convenience. The battery provides the necessary voltage required for the Holter monitor components to function.

With a battery capacity of 600mAh, the energy capacity is given as;

$$\text{Total Energy Capacity} = \text{total battery capacity (mAh)} \times \text{battery voltage (V)}$$

$$\text{Total Energy capacity} = 600\text{mAh} \times 9\text{V} = 5{,}400\text{mWh}$$

Since the main power-consuming components are the AD8232 module, the ESP32 microcontroller, and a display module, which are all low-power modules, we can assume a power consumption of around 150mW.

$$\text{The Battery Life(hours)} = \frac{\text{Total Energy Capacity (mWh)}}{\text{Power Consumption(mW)}}$$

$$\text{The Battery life} = \frac{5{,}400\text{ mWh}}{150\text{mW}} = 36\text{ h}$$

AD8232

Bio-signals are electrical signals that are generated by living cells. They are characterized by low amplitudes and a variety of noises(Kaniusas, 2012). These bio-signals are recorded by electrodes attached to the user's skin. Due to their low amplitude and noise, they must be amplified and filtered to improve their quality, making them easy to analyze.

AD8232 is a front-end integrated circuit capable of amplifying and filtering biopotential signals from electrodes placed on the patient's skin. It is perfect for the design of wearables due to its low energy consumption (about 170 A), small size, fast restore function, wide temperature range, amplification, and filtering projects all on a single chip (Analog Devices, 2013; Eliseichev et al., 2022).

In this design, the heart's electrical signals were captured by the electrodes attached to the body. These captured signals were amplified and filtered by the AD8232 IC to produce accurate heart rate readings. To compensate for the relatively low millivolt levels of ECG signals, the AD8232 incorporates a three-stage amplifier into its signal-boosting process.

A low-noise instrumentation amplifier with a gain of 100 makes up the first stage. As shown in Figure 14, it consists of two transconductance amplifiers (GM1 and GM2) that are perfectly matched, a DC blocking amplifier (HPA), and an integrator made up of C15 and an op amp(Analog Devices, 2013). This is connected internally to a high-pass filter with capacitors and resistors with a cut-off frequency of 0.05Hz that can be calculated with the formula:

$$f_c = \frac{10}{2\pi\sqrt{R_{24}R_{25}C_{14}C_{15}}} \qquad \text{Equation 1}$$

But $R_{24} = R_{25} = 10\text{M}\Omega$, and $C_{14} = C_{15} = 3.2\mu\text{F}$

The formula can be simplified as:

$$\mathbf{f_c = \frac{10}{2\pi RC}} \qquad \text{Equation 2}$$

and the cut of frequency is calculated as follows:

$$f_c = \frac{10}{2 \times \frac{22}{7}(10\text{M}\Omega \times 3.2\mu\text{F})} = \frac{10}{201.143} = 0.049\text{Hz} \approx \mathbf{0.05Hz}$$

$$R_{21} = 0.14 \times R_{25} \qquad \text{Equation 3}$$

Where R_{24}, R_{25}, R_{21}, C_{14} and C_{15} are resistors and capacitors specified in Figure 13. R_{21} compensates for the second-order topology for the high pass to adjust the filter. The high pass filter eliminates baseline wander and removes low-frequency noise.

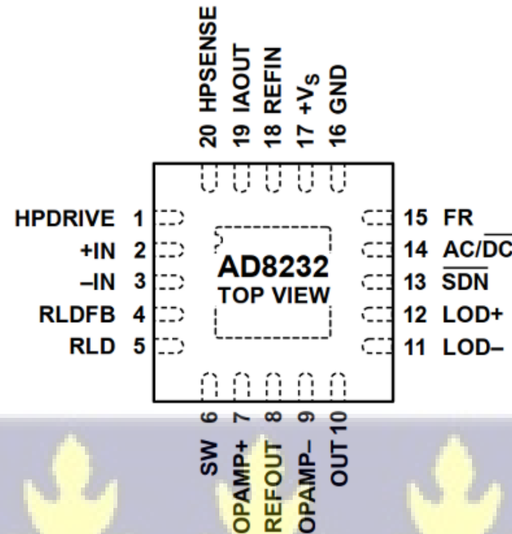


Figure 12. AD8232 Pin Configuration [74].

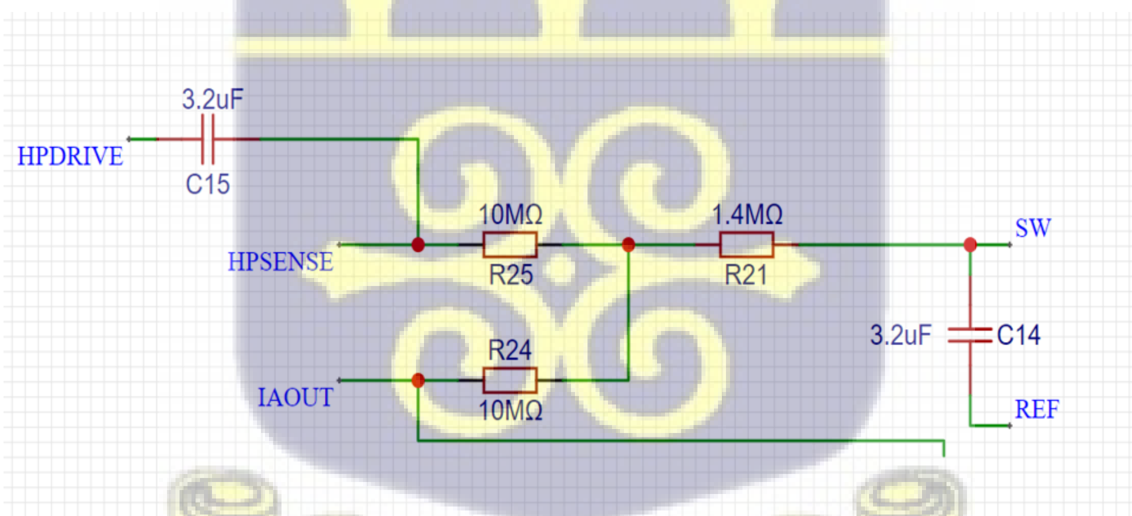


Figure 13. High Pass Filter (Researcher's own construct, 2023).

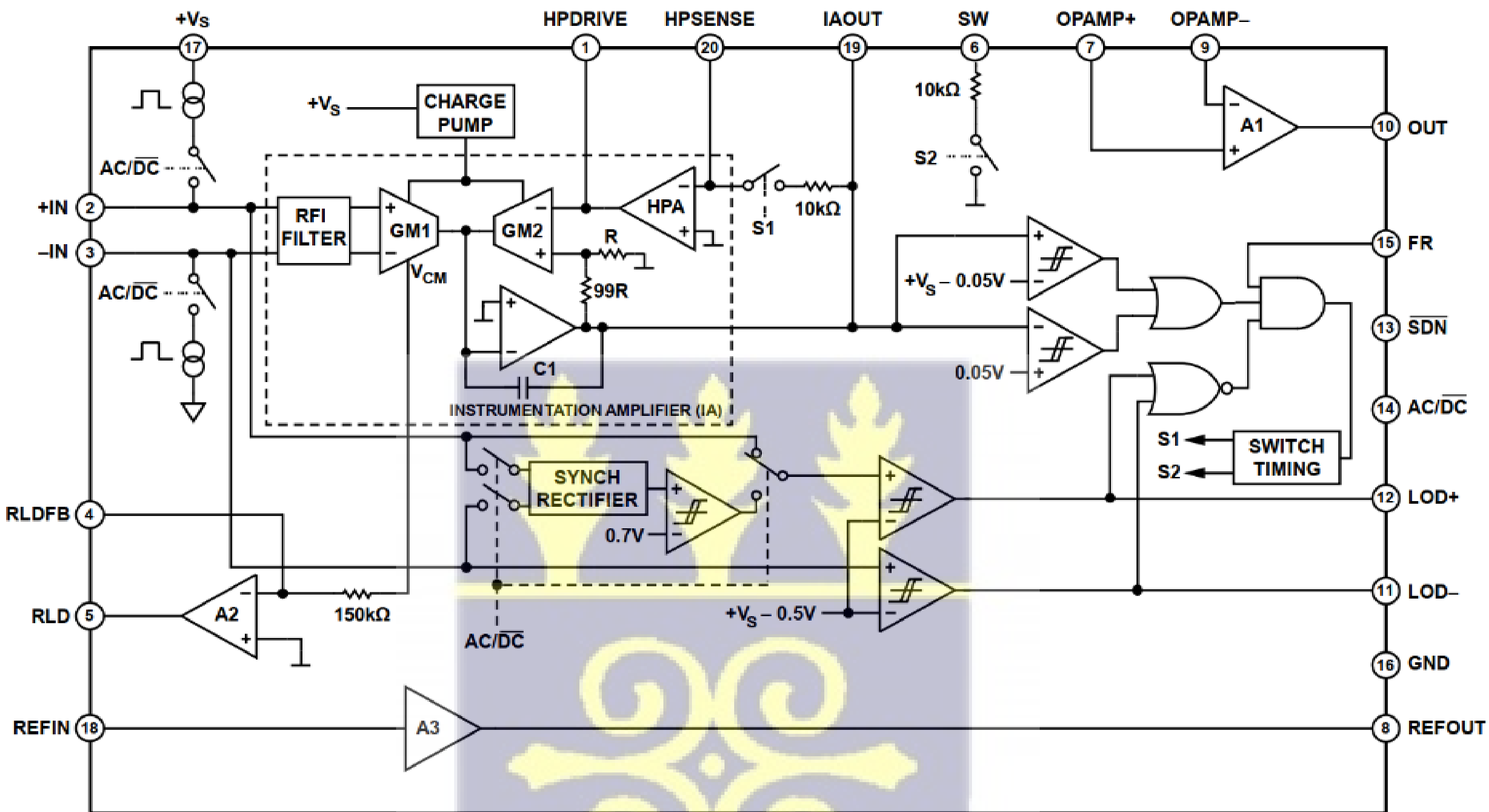


Figure 14. Simplified Schematic Diagram of AD8232 [74].

The second stage consists of a non-inverting amplifier with variable gain, which is determined by the ratio between the feedback resistor connected to the output of the first stage and the input resistor connected to the non-inverting input of the second stage. It is used to delimit the band of the desired biopotentials and attenuate the high frequencies. It can be set to have a gain of 1000, 100, or 10, which is enough to amplify the weak biopotential signals to a level that an ADC or a microcontroller can easily measure. The low-pass filter removes high-frequency noise and artifacts. The cut of frequency and the gain of the Holter monitor were calculated with the following equations:

$$f_c = \frac{1}{2\pi\sqrt{R_{32}R_{33}C_{19}C_{21}}}$$ Equation 4

but $\mathrm{R}_{32} = \mathrm{R}_{33} = 180\mathrm{K}\Omega$,

$\mathrm{C}_{19} = 22\mathrm{nF}$ and $\mathrm{C}_{21} = 1.5nF$

$$\mathrm{f_c} = \frac{1}{2\times\frac{22}{7}\sqrt{(180\mathrm{K}\Omega\times180\mathrm{K}\Omega\times22\mathrm{nF}\times1.5\mathrm{nF})}} = \frac{1}{0.0065}$$

$$\mathrm{f_c} = 153.85\mathrm{Hz} \approx \mathbf{150.\,Hz}$$

$$\boldsymbol{Gain} = \mathbf{1} + {}^{\boldsymbol{R_{34}}}/_{\boldsymbol{R_{31}}}$$ Equation 5

$$Gain = 1 + \frac{1M\Omega}{100K\Omega} = 11$$

Where $R_{31}, R_{32}, R_{33}, R_{34}, C_{19}$ and C_{20} are resistors and capacitors.

The third amplification stage is a crucial component of the AD8232. It is a non-inverting amplifier with a gain of 2, 4, or 6 and it is employed to further boost the amplifier's gain and increase its bandwidth.

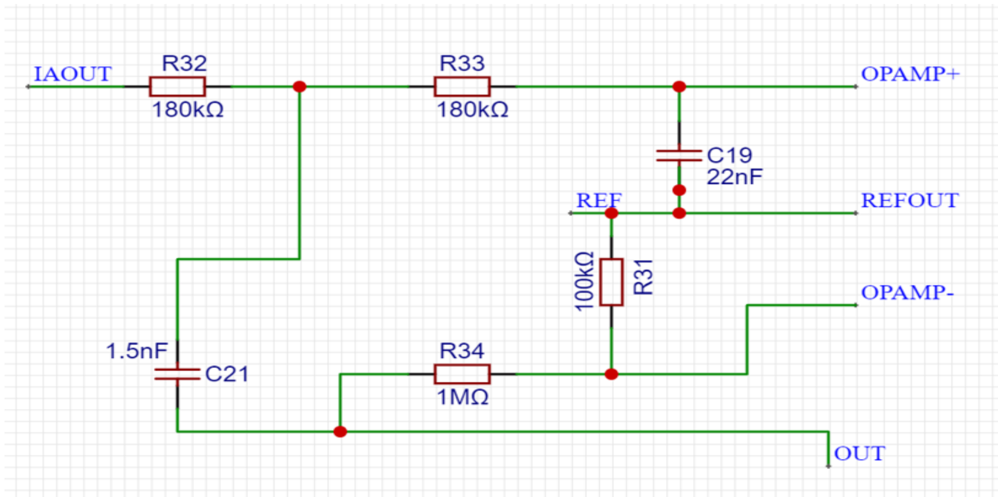


Figure 15. Schematic diagram of the Low Pass Filter (Researcher's own construct, 2023).

ESP32

A microcontroller is a small, programmable computer built on a single chip that controls electronic devices. The Espressif ESP32 microcontroller is a versatile and powerful hardware platform for implementing Internet of Things (IoT) applications. It is the successor to the ESP8266 microcontroller that was very popular thanks to its low cost and high functionality. ESP32 microcontroller can be viewed as an "ESP8266 on steroids," as it has higher processing power and more inputs and outputs. It is a dual-core 32-bit microcontroller with up to 240 MHz clock frequency and can connect to Wi-Fi as well as Bluetooth Low Energy (Foltýnek et al., 2019). Besides the ability to connect to the internet, it has a wide range of peripherals such as ADC, DAC, UART, I2C, SPI, and PWM. The ESP32 also has very low power consumption when not powered on, and when using sleep modes, it can achieve between tens of microamperes and a few hundred microamperes. The microcontroller in this study is responsible for acquiring the analog signals from the AD8232 sensor and processing them to obtain quality signals for the accurate detection of arrhythmia. Again, it controls how the processed data is stored, whether it be on an external memory card or transmission to the cloud storage. The resulting architecture allows for very efficient data processing and transmission, making it helpful for remote monitoring (Babiuch et al., 2019).

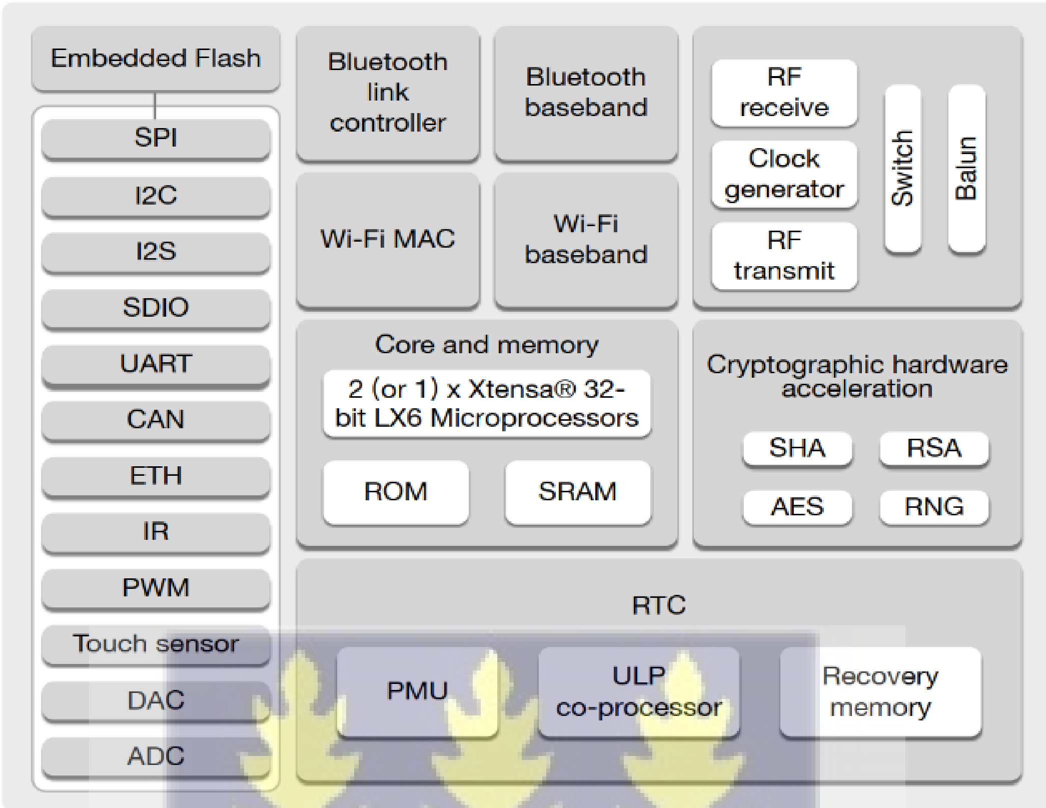


Figure 16. Functional Block Diagram of ESP32 (Foltýnek et al., 2019).

SD Card Module

The Holter monitor uses the SD card module to store data. It connects to an SD card, which gives it non-volatile memory for storing the ECG data that has been recorded. With the SD card module installed, the Holter monitor can store extensive amounts of ECG data for an exceptionally long period. This enables medical professionals to review and analyze the recorded data at a later time for precise diagnosis and treatment advice. It offers a dependable and portable storage solution, making data transfer and archiving simple.

OLED display

The SSD1306 is a popular and widely available display driver chip that can be used with a variety of microcontrollers and platforms. It is designed for a standard OLED panel with a cathode. The SSD1306 integrates contrast control, display RAM, and an oscillator, thus minimizing the requirement for external components and reducing power consumption(SOLOMON SYSTECH, 2012). It is used in this design to display the heart-beat at a point in time.

Circuit Schematic and Printed Circuit Board (PCB) Design

The design of the Holter monitor PCB demanded a systematic methodology to ensure precision and reliability. In this project, EasyEDA which is a web-based electronic design automation (EDA) tool(Access, 2021) was used to design the electronic schematic board and PCB. It offers a user-friendly environment for the initiation, design, and realization of complex PCB layouts. EasyEDA provides a wide selection of symbols and components that are easy to add to the schematic.

The design of the PCB involved three main steps; the schematic design, PCB layout design, and manufacturing processes.

Schematic Design Process: Important components such as the ESP32 chip, SD card, AD8232 heart sensor connections, buzzer, OLED screen, and power supply formed the basics for the schematic design. These components were placed on an interface provided by EasyEDA and were connected through wiring to accurately represent the electronic circuitry of the Holter monitor. The various components of the circuit and how they are connected are shown in Figure 17.

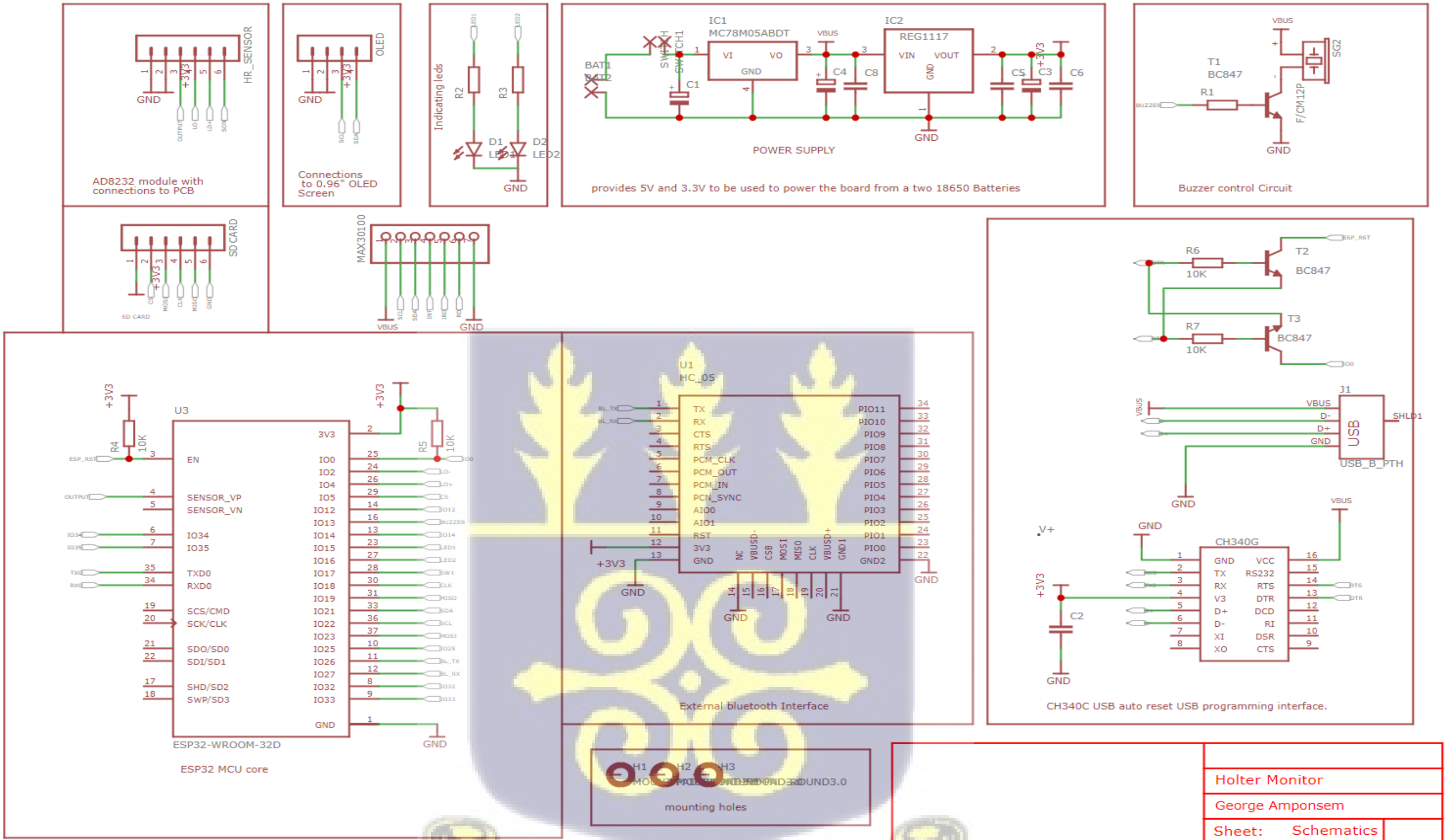


Figure 17. Schematic Diagram of the Holter Monitor. It shows the interconnection of the AD8232 ECG sensor, ESP32 microcontroller, Bluetooth chip, USB port, USB bridge, auto reset circuit, buzzer, SD card, and other supporting components to create a working circuit. It gives a general overview of the power distribution, signal paths, and electrical connections inside the Holter monitor (Researcher's own construct, 2023).

PCB Layout Design Process: The printed circuit board was designed by transitioning the schematic diagram in Figure 17 into the PCB layout. In this process, careful considerations were given to the arrangement of components on the board, taking into account spatial constraints and ensuring an efficient flow of signals. The batteries were mirrored so that it goes to the back of the PCB. The AD8232 is situated close to the microcontroller to eliminate noise. Again, a 100nF decoupling capacitor was placed nearby to further filter out noise. After laying out the PCB, traces were drawn to enable the transmission of electrical signals between various parts and signal integrity. The PCB layout guarantees proper power distribution, signal routing, and mechanical stability of the Holter monitor as shown in Figure 18.

EasyEDA also offers a 2D visualization feature that enables a representation of the PCB design (see Figure 19). This shows how the components are arranged in two dimensions. It offers a virtual representation of the Holter monitor, which makes it possible to gain a deeper comprehension of the overall factor, component placement, and utilization of space. The 2D view provides a visual understanding of how the PCB and its parts work together to form the finished product.

Manufacturing Process: After a systematic review of the PCB layout was done to ensure concise placement of the components and accurate trace routing, Gerber files were generated within EasyEDA. These files contain the necessary information for PCB manufacturing. The Gerber files were then sent to JLCPCB, a reliable PCB manufacturing company with a stellar reputation for accuracy and quality to help us materialize our design. They took the PCB layout that was designed and turned it into a physical board that would act as the Holter monitor's framework.

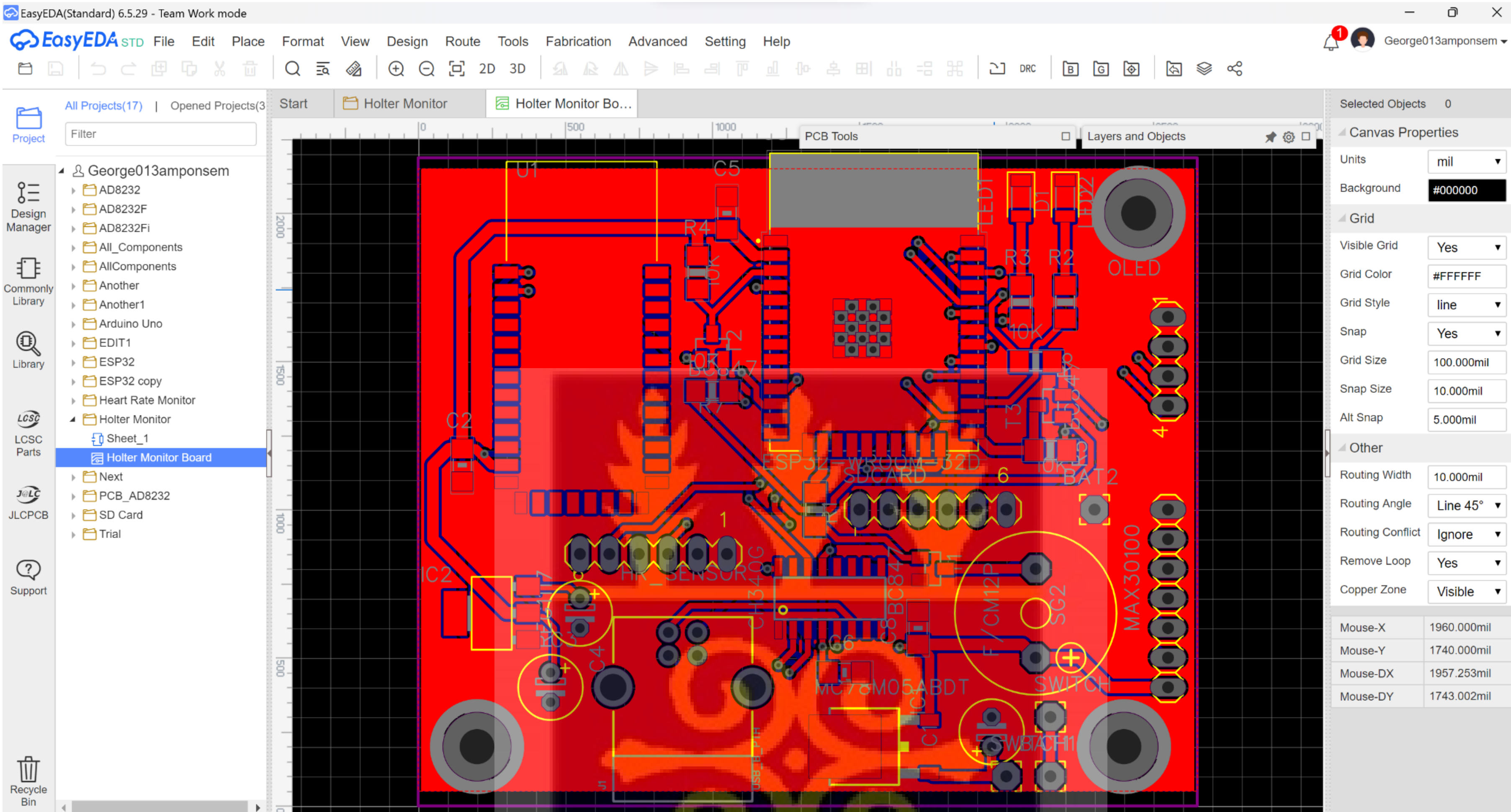


Figure 18. Printed Circuit Board Layout Design. This shows the layout and arrangement of the components on the PCB describing where the ESP32 chip, SD card, AD8232 ECG sensor, and other important parts are located on the circuit board. It draws attention to the connections and traces that enable the transmission of electrical signals between various parts (Researcher's own construct, 2023).

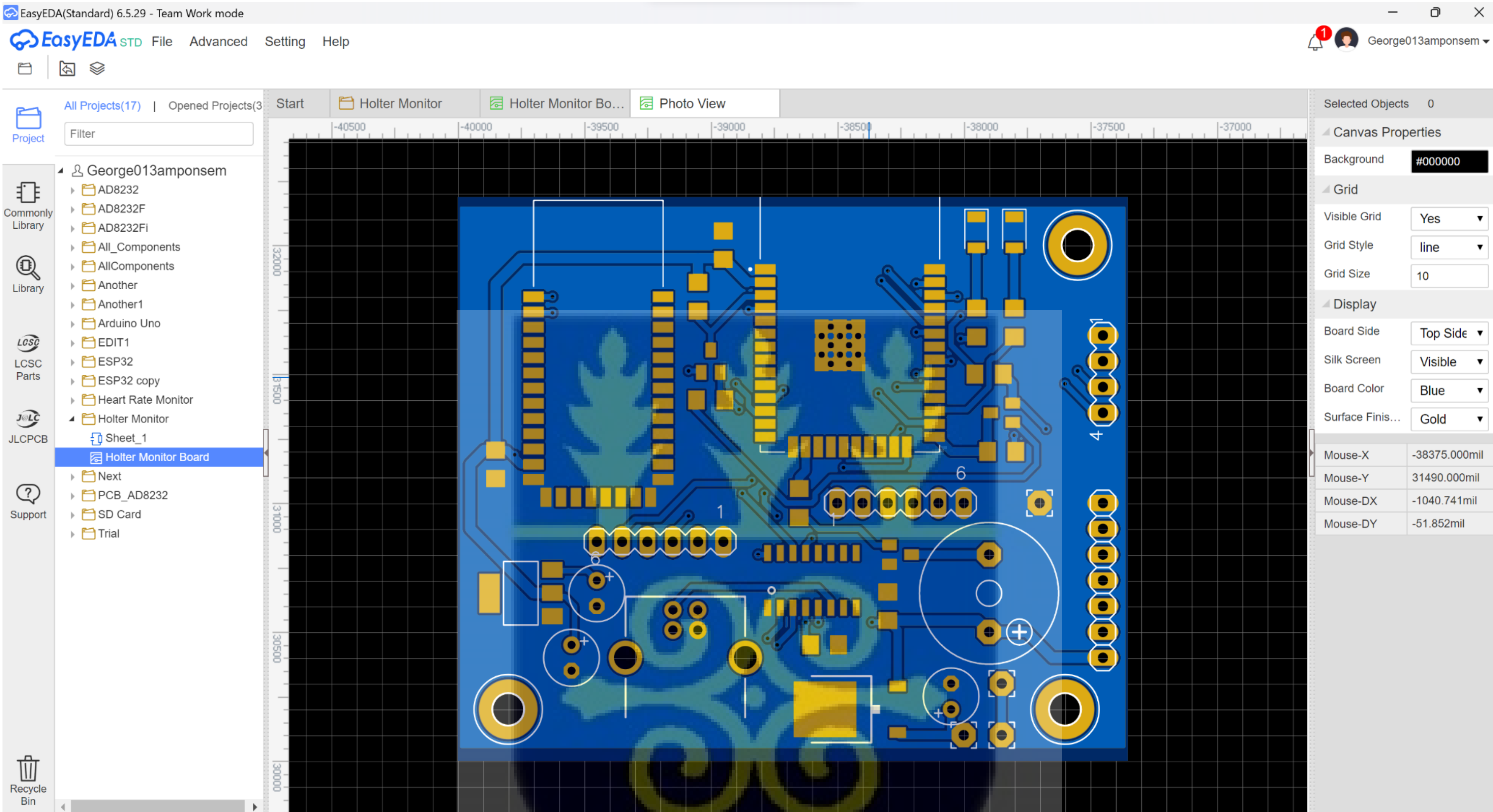


Figure 19. Front View of the 2D model of the Printed Circuit Board. This is a representation of how the actual board will look like (Researcher's own construct, 2023).

PCB Assembly

The fabricated board was then shipped to me after two (2) weeks. After receiving the newly manufactured PCB (see Figure 20), the next stage involved the complex process of assembling the components. A vacuum chuck was used to hold the board securely in position during component assembly. This was done to ensure that the soldering process was steady and accurate. To aid in the smooth distribution of soldering paste, the board was prewarmed using the soldering station. Soldering paste was then spread evenly across the board using a dispensing tool. The paste contains flux and solder particles that facilitate the soldering process.

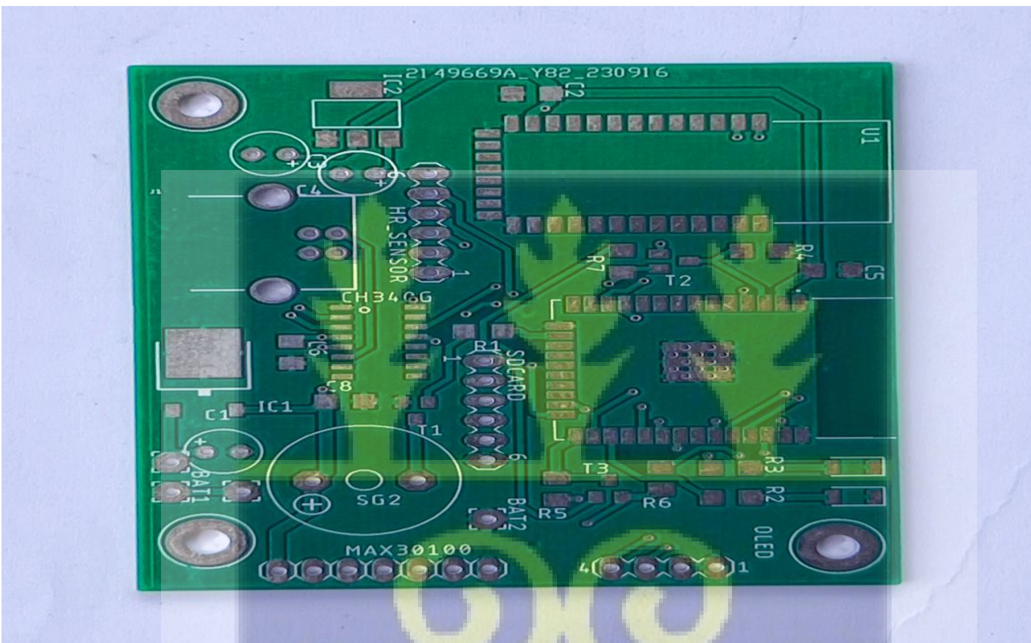


Figure 20. Printed Circuit Board from JLCPCB (Researcher's own construct, 2023).

The surface-mount components, which have small soldering pads, were carefully positioned on the board using a tweezer. The PCB was then placed on a hot plate, and the temperature was set to 230 degrees Celsius. The hot plate heats the board evenly, melting the solder paste and creating strong solder joints between the components and the board. After the soldering was complete, the board was allowed some time to cool down. The remaining components (through-hole components) were manually soldered onto the board using a soldering iron and solder wire.

To fulfill the requirements of the project, the firmware on the board must be able to detect the user's heart rate and rhythm and then either store the recorded data on an SD card or transmit it wirelessly onto cloud storage using Wi-Fi to be accessed by a smartphone application. The firmware was developed with the Arduino IDE, which is an open-source platform designed for fast prototyping

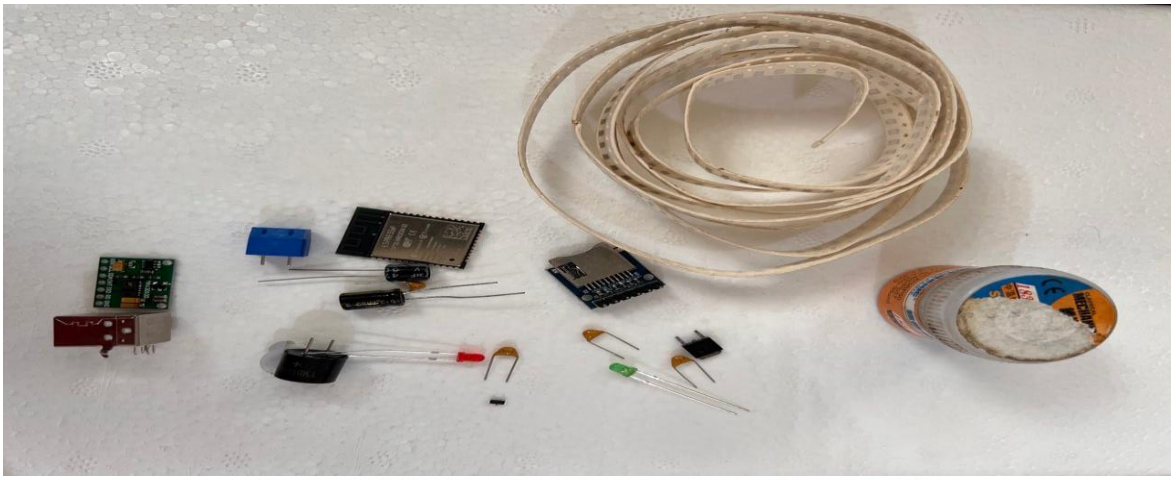


Figure 21. The figure shows the soldering paste, and the various surface mount and through hole used the assembly stage (Researcher's own construct, 2023).

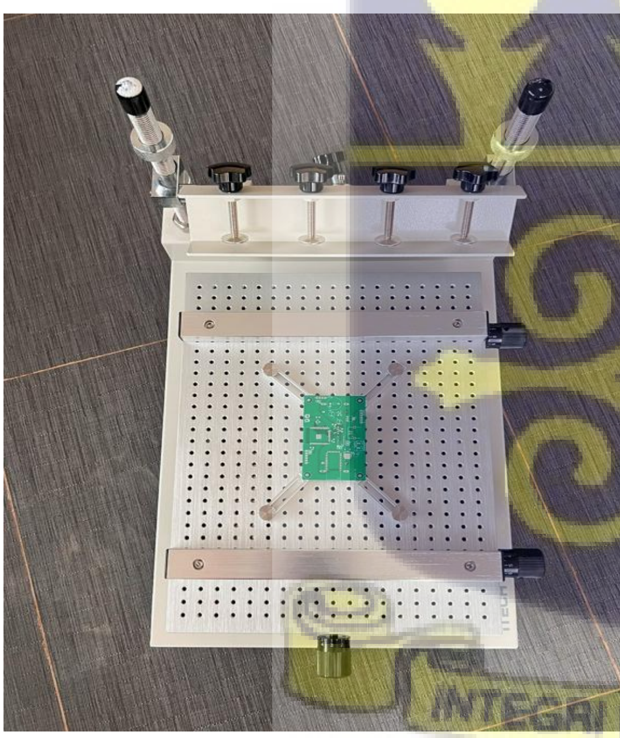


Figure 23. Vacuum Chuck securely holding the printed circuit board in place before soldering paste is spread on it(Researcher's own construct, 2023).

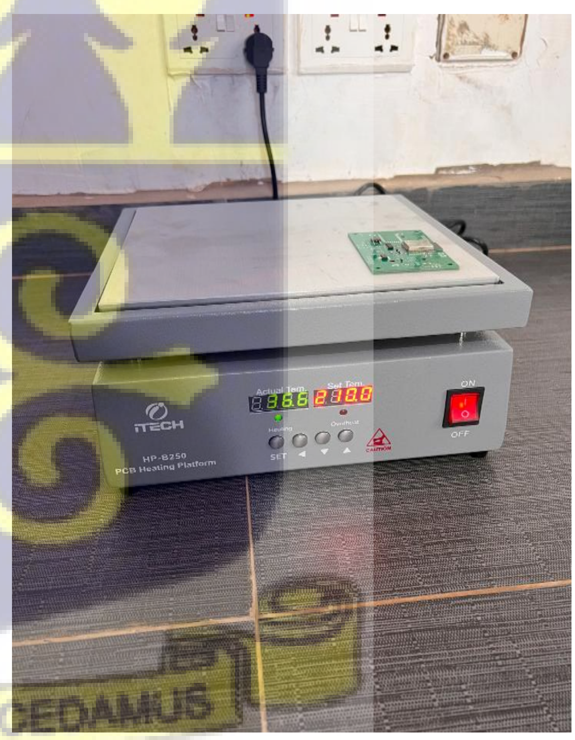


Figure 22. The Printed Circuit Board placed on a hot plate with temperature set to 230 degrees Celsius to melt the solder on the board to secure the solder joints (Researcher's own construct, 2023).

3.1.9.2 Enclosure and Wearable Design

It was vital to design a non-invasive device that would protect the user's body from experiencing any discomfort or irritation. For this reason, the electronic module, which contains the circuit board and a battery is encased in an acrylic material that is durable and rigid. It was manufactured using laser cutting technology, which was thought to be the fastest and most effective process. Laser cutting is the precise cutting of materials using a focused laser beam(Naresh & Pankaj, 2022). LaseCAD (version 8.18.6) was used to design the layout of the enclosure (see Appendix E) with a dimension of 64mm x 31mm x 74mm. A boss laser cutting machine at Ahsesi University was used to cut out the enclosure for the board. The dimensions of the enclosure were copied onto the machine, and with extreme accuracy and precision, it cut the enclosure. Precision was essential to making sure the components fit tightly and were effectively shielded by the container. The electronic module contributes to the Holter monitor's overall durability and dependability. The performance and lifespan of the Holter are improved by making sure the essential electronics are well protected, which also lowers the possibility of failure due to damage.

The wearable module consists of a cotton pouch and fasteners. It ensures the electrodes are securely attached to the user's body to capture high-quality signals and also protects the electronics enclosure, both of which are critical functions of the wearable Holter monitor system. Using cotton, a carefully chosen biocompatible material, emphasized the module's dedication to user comfort during extended wear. The fasteners used in the design also permit adjustments to be made to accommodate various body shapes and sizes. In addition, the cotton fabric helps to regulate moisture, creating a pleasant and dry atmosphere surrounding the electrodes.

3.1.9.3 Electrodes

Three silver/silver chloride (Ag/AgCl) disposable electrodes from Trace 1 were used to obtain ECG signals from the user's skin. These electrodes have excellent conductivity and stability. Although gold electrodes provide excellent conductivity and durability, their higher cost tends to render them less economically viable for disposable applications. Similarly, graphite electrodes, despite their low cost, exhibit poorer stability and signal quality in comparison to Ag/AgCl electrodes. Adopting Ag/AgCl electrodes was mostly due to their high conductivity, reliability, low polarization, and low cost. The fact that they are disposable makes them hygienic and reduces the need for cleaning or maintenance, which in turn minimizes the possibility of cross-contamination between patients.

The Ag/AgCl layer gives a stable reference potential for accurate ECG readings. They can be repositioned accurately when moved without having to be replaced. Incorrect ECG lead placement can lead to an increase in noise and artifacts, incorrect diagnoses, and other problems (Medani et al., 2017). In comparison to other electrodes used in ECG signal acquisition, Ag/AgCl is stable, has minimal polarization, and is readily available, making it a dependable and cost-effective method of obtaining high-quality ECG signals.
Each electrode is color-coded and has a unique lead name for quick identification. The AD8232 module receives the ECG signals for amplification and filtering from the electrodes through a lead wire.

Wearable Hardware Placement on Body

Standard ECG monitoring employs 12 leads, which offer multiple perspectives of the heart's electrical activity and allow for the identification and analysis of a variety of heart conditions. As the number of leads influences the placement of electrodes on the patient's body, implementing a

Figure 25. Disposable Ag/AgCl electrodes from Trace 1 used in this project (Researcher's own construct, 2023).

similar electrode configuration would make the wearable device too bulky and uncomfortable to be worn continuously.

We used a Lead II ECG placement as per the specifications of this project to ensure patient comfort while maintaining adequate ECG quality, as shown in Figure 24 below (Parveiz et al., 2023). The Lead II setup involves the use of three electrodes, where the left leg (LL) serves as the positive electrode, the upper arm (RA) serves as the negative electrode, and the right leg (RL) serves as the reference electrode (RL).

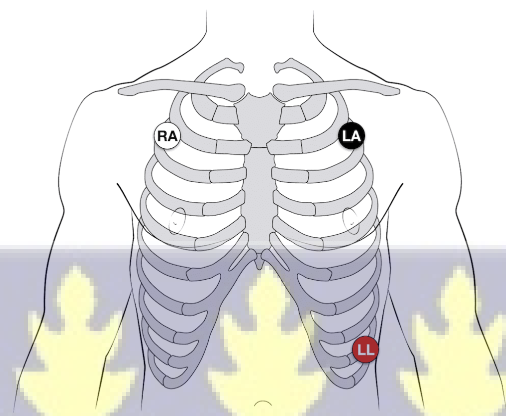


Figure 24. Lead II Electrode Placement (Sandau et al., 2017).

This specific placement is based on the Einthoven triangle (MANN, 1920) (see Figure 25). Einthoven's triangle uses vector summation to interpret the small electrical impulses that spread directionally as a result of the contraction and relaxation of the heart by placing electrodes on the limbs. It records changes in electrical potential, showing how the heart chambers depolarize and repolarize with each beat. By placing the positive electrode on the left leg and the combined right and left arm as the negative electrode, a closed circuit for heart activity is formed. This setup detects electrical signals from various perspectives, focusing on the heart's horizontal plane, enabling effective assessment of core cardiac rhythm(Kalra et al., 2019; Pura, 2022). Although it is not as extensive as the 12-lead configuration, the Lead II arrangement is capable of detecting most arrhythmias, ischemia, and infarction, making it suitable for continuous ambulatory monitoring(Sandau et al., 2017).

Since it is impossible to standardize electrode placement and orientation for all subjects, the electrodes are equipped with long cables that can be comfortably stretched to any desired location. Before attaching the electrodes, the skin is prepared by cleaning it with an alcohol swab to remove oils and dirt and if there is excessive hair in the electrode placement areas, they may be shaved or trimmed to ensure better contact between the electrodes and the skin. This preparation helps minimize impedance and improves the electrode-skin contact (Litfl Team, n.d.).

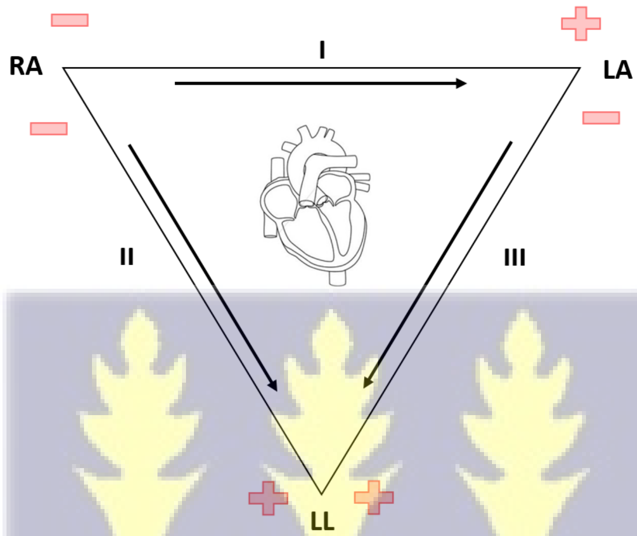


Figure 25. Einthoven's Triangle showing the three limb electrodes, I, II and III forming a triangle at the right arm (RA), left arm (LA) and left leg (LL) (Litfl Team, n.d.).

3.2 PHASE II: SOFTWARE DEVELOPMENT

This section describes the methods adopted in the design and development of the software components of the wearable Holter monitoring system. The software subsystem comprises embedded firmware to extract, preprocess, store, and transmit ECG signals, and a mobile application for data visualization, storage and user interaction.

3.2.1 Design and Development Approach

Agile methodology was used for the software development of this project. It is a project management technique that entails dividing the project into distinct phases and prioritizes continuous collaboration and progress. It has been demonstrated that this project management approach is more effective than the conventional waterfall approach. It encourages an iterative and incremental development approach, in which the app's functionality is added in small steps. This allows for continuous improvement based on user input and evolving requirements by allowing for early releases and quick feedback. The design process is summarized below(iXora Solution, n.d.):

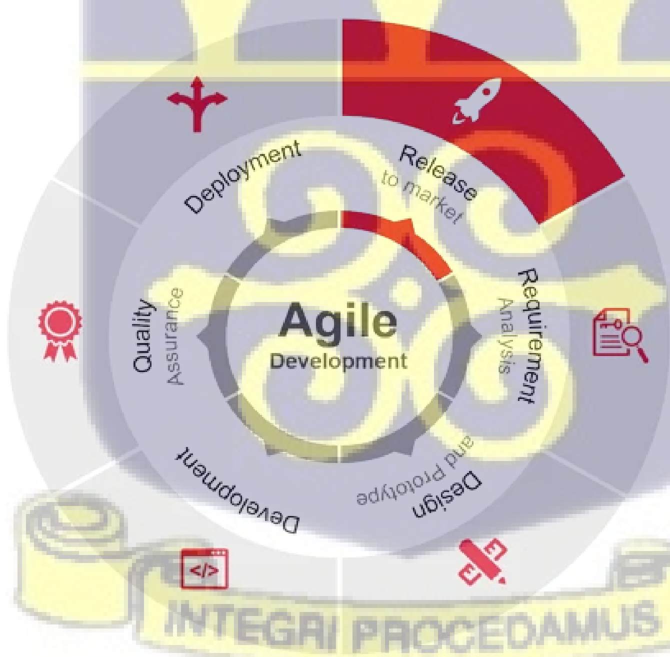


Figure 26. Agile Project Management Workflow (iXora Solution, n.d.).

3.2.2 Embedded Firmware Development

Firmware is primordial code that is programmed into the microcontroller. It translates the design of a project into a language usable by the hardware components (Saiz-Vela et al., 2020). This section provides a comprehensive study of the firmware, highlighting key components, libraries, and features that support the monitoring of cardiac activity.

Arduino Integrated Development Environment (IDE) is an open-source software application that enables the writing and uploading of computer programs or firmware to Arduino-compatible boards. It provides a cheap way to implement both simple and complex projects across a wide range of hardware platforms (Badamasi, 2014). The development environment is easy to use with a graphical user interface (GUI) that allows for a straightforward project development workflow from very basic tasks such as writing code snippets to complex tasks involving a wide use of various libraries.

In the development of the firmware, the Arduino IDE version 2.1.0 was the primary platform for the integration of our code that define the behavior of the Holter Monitor. It establishes a Wi-Fi connection to transmit heart rate data in real-time using MQTT protocol to a broker hosted at broker.hivemq.com. The heart rate and rhythm are published to specific topics, allowing remote monitoring. The firmware also includes initialization routines for an OLED display and SD card module which allows for local real-time visualization of the heart rate data and secure data logging for later analysis. This ensures that even in the absence of an internet connection, data is preserved and reviewed.

In operation, the firmware samples heart rate data at regular intervals, processes this data to calculate the pulse rate, and detects the heart rhythm. The results are displayed on the OLED screen and logged to the SD card. Alerts are managed through LEDs and a buzzer, which provides immediate feedback if abnormal heart rates ae detected.

Essentially, the Arduino IDE serves as our conductor, directing all of the components to execute their respective functions in unison.

3.2.3 Mobile Application Development

In this phase of the project, the objective was to design and develop a smartphone application that wirelessly communicates with wearable hardware. The app's functionalities include receiving and securely storing real-time streamed data of ECG and heart rate.

3.2.4 Software Requirement Analysis

The requirements for the smartphone application have been categorized into non-functional and functional requirements.

Non-Functional Requirements

Non-functional requirements are the criteria used to evaluate a system's operation, focusing on aspects beyond its core functionality. These requirements address key performance and operational aspects, ensuring the system's overall quality. In the context of the Holter Monitor, non-functional requirements like security, safety, ergonomics, and convenience are crucial for providing a reliable, secure, and user-friendly experience.

Security measures, such as firewalls and secured databases, protect patient information and heart rate data from unauthorized access and breaches. Safety ensures that the system handles and processes data correctly and reliably, minimizing the risk of failure or harm. Ergonomics focuses on creating a user-friendly interface, effective communication, and a simple registration process, enhancing user interaction and satisfaction. Convenience includes visualization of data (heart rate and ECG graph) and aesthetics, contributing to a positive user experience.

By prioritizing these non-functional requirements, the Holter Monitor can minimize failures, ensure uninterrupted service, and increase user trust. It is essential to consider non-functional requirements throughout the development process to deliver a product that not only functions correctly but also perform well, is secure, reliable, maintainable, and provides a positive user experience. Neglecting these requirements can result in user dissatisfaction, system failures, security breaches, and increased operational costs.

Functional Requirements

Functional requirements outline the specific actions or tasks that a system must perform to meet the needs of its users. These requirements describe the desired behaviors of the system, focusing on what the system should do to satisfy user needs.

In the context of the Holter Monitor, functional requirements include securing a database for storing patient information and heart rate data. The system must provide alerts for abnormal heart rates and ensure real-time data transmission. Additionally, the Holter Monitor must support compatibility with heart rate and ECG graph visualization.

Functional requirements play a vital role in defining the fundamental capabilities and behaviors of the Holter Monitor. They serve as the foundation for development, testing, and validation, ensuring that the final product aligns with user expectations and business objectives. To ensure their implementation is correct, functional requirements are typically tested and verified, and they should be technically feasible and aligned with the project's scope.

Below is an objective tree highlighting all the system requirements.

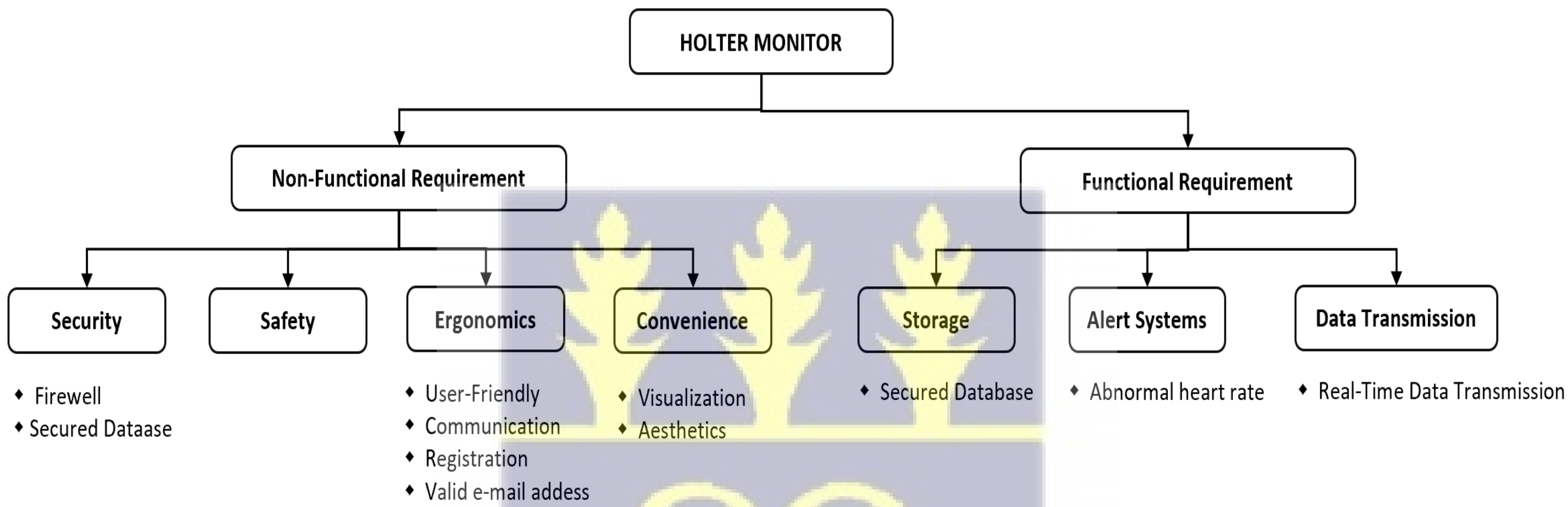


Figure 27. Mobile Application Requirement (Researcher's own construct).

3.2.5 Software Specifications

Specification	Description
Programming Tools	Gradle, MPAndroidChart
Programming Language	Java
Integrated Development Environment (IDE)	Android Studio

Figure 28: Mobile Software Specifications (Researcher's own construct).

3.2.5.1 System Development and Programming Language Selection

The process began with the creation of a basic model based on an initial set of specifications. The model was then tested, and feedback was collected. This feedback was used to refine the project's specifications, and the model was extended accordingly. This cycle of testing, feedback, and refinement was repeated until all specifications were met, resulting in a fully functional software application.

The software application was developed for Android (can be accessed at (https://github.com/George-dot/Holter_Monitor-App). The Android application relies on a blend of native Android SDK tools and third-party libraries to enable real-time ECG monitoring. By using Android Studio, Java, and a range of libraries for signal processing, connectivity, user interface, and background processing, the application is designed to be robust, responsive, and capable of handling real-time data processing and analysis. Below are the various development tools and programming languages used:

1. Programming Tools

The Tech Stack for developing the front end of mobile applications was identified and these are:

- Gradle: It is the build automation tool integrated into Android Studio, responsible for managing project build configurations, dependencies, and automating tasks. In an Android project, the build gradle file plays a crucial role in defining project dependencies, such as libraries, build types, and custom build tasks. By automating routine tasks and ensuring consistent builds, Gradle simplifies the development process, saving time and effort.

- MPAndroidChart: It is a versatile and feature-rich charting library designed specifically for Android applications. It offers a wide range of chart types, including line charts, bar charts, pie charts, and more, along with extensive customization options. In the context of an ECG monitoring app, MPAndroidChart can be leveraged to create dynamic, real-time visualizations of ECG data, providing users with a clear and intuitive way to monitor their heart's electrical activity.
- Material Components: Material Components for Android bring Google's Material Design principles to life, providing a unified design system that spans visual, motion, and interaction design across multiple platforms. By incorporating Material Components, the app can boast a sleek, modern, and intuitive interface that adheres to a consistent design language, ultimately enhancing the user's experience.

***2.* Programming Language**

- Java: Java is a fundamental programming language in Android development, renowned for its platform independence, rich library support, and vast community resources. The language's object-oriented design and extensive API make it an ideal choice for building sophisticated applications, such as real-time ECG monitoring systems, which require complex logic and precise control(Brito et al., 2018).

***3.* Integrated Development Environment**

- Android Studio: It is the officially recognized Integrated Development Environment (IDE) for Android app development. It offers a wide range of tools that facilitate the development, testing, and debugging of Android applications. One of its key features is the ability to test apps on a diverse array of virtual devices with varying configurations. Additionally, Android Studio comes equipped with an integrated build system that simplifies dependency management and automates tasks, streamlining the development process.

3.2.6 Software System Design

The software system design integrates several components, including the MQTT, communication framework, Wi-Fi connectivity, database management, signal preprocessing, and user interface modules. Figure 29 shows an overview of the smartphone application architecture. Each part of the software architecture is explained in detail in the following sections:

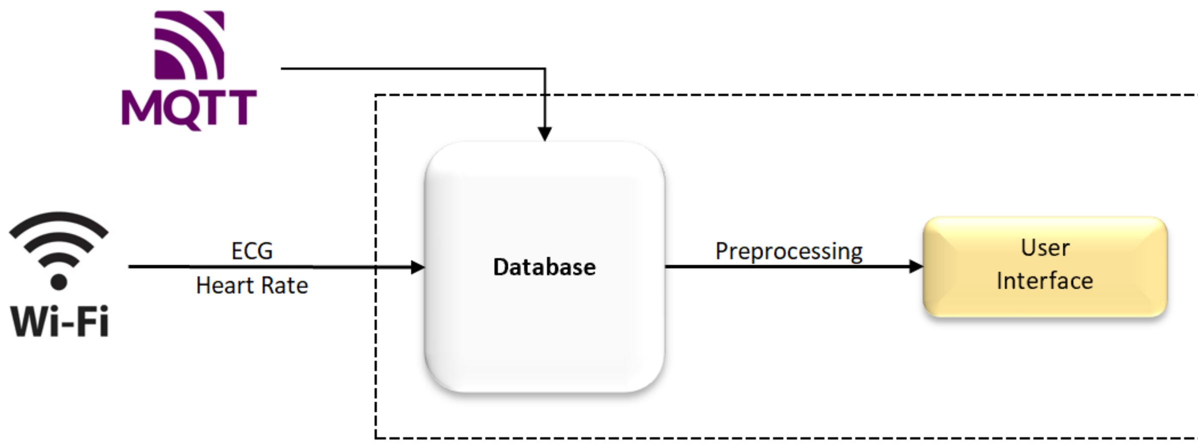


Figure 29. Overview of the smart phone application architecture (Researcher's own construct).

MQTT Protocol

MQTT (Message Queuing Telemetry Transport) is a lightweight messaging protocol designed for efficient communication in networks with limited bandwidth or high latency. The system operates based on a publish-subscribe concept, which separates the message sender (publisher) from the message receiver (subscriber). The essential component of MQTT is the broker, which acts as a server that receives messages from publishers and directs them to the relevant subscribers based on a hierarchical topic structure. Publishers send messages labeled with topics to the broker without concern for the recipients, while subscribers register their interest in specific topics with the broker. MQTT offers various degrees of Quality of Service (QoS) to guarantee the dependability of message transmission. It also includes features such as retained messages and the "Last Will and Testament" (LWT) to enhance communication resilience (Hunkeler et al., 2008; Shanmugapriya et al., 2021). In the mobile application, MQTT is used to facilitate efficient and reliable communication between the Holter monitor and the mobile device. The Holter monitor acts as a publisher, sending ECG and heart rate data to the broker, to which the mobile application subscribes.

Wi-Fi Module

The Wi-Fi module serves as the intermediary connecting the mobile application to the MQTT server. It enables the ECG data to be transmitted wirelessly to the mobile device. The use of Wi-

Fi ensures real-time data transfer, allowing continuous monitoring without the need for physical connections.

Database

The database component of the mobile app architecture serves as a repository for all the users' information. It is responsible for storing and managing various types of data generated or used by the application, such as user profiles and authentication information, ECG and heart rate readings collected over time, user activity logs, settings, and preferences. The database uses NoSQL to store and retrieve data efficiently. Access restrictions are implemented as security measures to protect sensitive health-related information.

Data Preprocessing

Preprocessing in the context of ECG signal processing refers to a sequence of processes aimed at refining the raw ECG data and making it ready for analysis. In this study, the signals were subjected to digital filters to eliminate noise and detect QRS, as suggested by Pan and Tompkins (Pan & Tompkins, 1985; Thurner et al., 2021). The raw signals acquired from the Holter monitor undergo filtration using a bandpass filter set between 0.5 Hz and 40 Hz to remove any unwanted noise. The resulting signal is subjected to differentiation to highlight rapid variations in the signal, facilitating the identification of the QRS complex; squared to amplify the peaks of the signal, facilitating the detection of R-waves; and integrated using a moving window to enhance the smoothness of the signal, further highlighting the QRS complex and improving its visibility.

User Interface (UI)

The main aim of the user interface is to display real-time graphing of the ECG signal and the heart rate. The app dynamically updates the UI and provides timely notifications based on the analyzed ECG data. To do this, it continuously refreshes the ECG graph to reflect the most recent payload data, ensuring the user has an up-to-date view of their heart activity.

It is also designed to instantly notify the user through Android's notification system if there is any heart rate out of the range set, ensuring prompt awareness of potential heart-related issues.
The user interface also includes a history tab that stores the wirelessly streamed data from the Holter monitor for later analysis.
Once the application is started;

1. The user is asked to log in or sign up. This step requires the user to have an email and also provide the date of birth of the user.
2. The next window displays the dashboard where the current heart rate, the heart rhythm, and a graph of the heartbeats are displayed. This interface also has numerous buttons including a history and a settings button.
3. The user is prompted to connect to the internet for data retrieval if there is no internet connection. When this is done, the status changes from "disconnected" to "connected."
4. The dashboard includes an "Activity Status" feature that allows the patient to select their current activity level. They can choose between different levels such as resting, engaging in normal aerobics, or performing hard aerobics to indicate their current physical activity.
5. The dashboard also has a "History" button. The history page has two tabs, "ECG" and "Heart Rate." Each tab activates a window, where the history of the heart rhythm and heart rates that fall outside the specified range based on the chosen activity status are stored respectively
6. Lastly, the Settings" page has a "Silence alarm" key that is used to silence the alarm whenever the heart beat falls outside the specified range.

3.2.7 Activity Recognition

The user's activity is determined by the information that they provide to the application. Users have the option to choose between three activity levels: "resting" for low-intensity activities such as sitting or lying down, "normal aerobics" for moderate-intensity activities like walking, and "hard aerobics" for high-intensity activities such as running or jogging. After the user selects the desired option, along with the age provided during the sign-up process, the system proceeds to compute the range of heartbeats using Equations 6 and 7. These equations were proposed by the American Heart Association (AHA) and historically use the Target Heart Rate calculation, which is based on the percentage of the maximum heart rate (Nes et al., 2013). Upon detecting heart rates outside the predetermined range, the mobile application activates an alarm, accompanied by a display message signaling either "heart rate too fast" or "heart rate too slow," based on whether the recorded heartbeat surpasses or falls below the calculated threshold.

$$THR = HR_{max} \times Intensity \qquad \text{Equation 6}$$

Where THR = Target Heart Rate and HRmax = Maximum Heart Rate.

The maximum heart rate may also be calculated using the formula below.

$$HR_{max} = 211 - 0.64 \times age \qquad \text{Equation 7}$$

This formula gives a more accurate estimate of the maximum heart rate by taking into account how the maximum heart rate changes with age.

Table 6. Various activities and intensities

Activity	Intensity	Description
Resting	40% - 50%	The user is idle (sitting or sleeping)
Normal Aerobics	50% - 70%	The user is walking
Hard Aerobics	70% - 85%	The user is running, jogging, etc.

3.3 PHASE 3: SIMULATION

The Holter monitor system was tested using a Fluke ECG simulator (ProSim 8) to evaluate its accuracy in capturing heart rate and heart rhythms. The Fluke ECG simulator generates a range of ECG waveforms, including both normal and abnormal heart rhythms (see Figure 30). Electrodes from the Holter monitor were connected to the corresponding pins on the simulator and interfaced with a laptop using a USB cable.

Figure 30. Fluke ECG Simulator (ProSim 8) with the electrode connections.

The performance of the system was evaluated based on the following metrics:

- Heart Rate Accuracy: The heart rate values computed by the wearable Holter system were compared with the reference heart rate values generated by the ECG simulator.
- Signal-to-Noise Ratio (SNR): The quality of the acquired ECG signals was assessed by analyzing the signal-to-noise ratio before and after signal processing.
- System Latency: Latency was measured as the time delay between ECG signal acquisition by the wearable device and visualization on the mobile application or computer interface.
- Arrhythmia Detection Capability: The system's ability to detect abnormal heart rate conditions was evaluated by configuring the ECG simulator to produce heart rates outside normal physiological ranges and observing the corresponding alerts generated by the system.

The results obtained from these evaluations are presented and discussed in Chapter Four.

CHAPTER FOUR

RESULTS AND DISCUSSION

This chapter presents the results obtained from the design and implementation of the proposed low-cost wearable Holter monitoring system. The results are organized in line with the project objectives to demonstrate the construction of the wearable device, the development of a mobile application for wireless data visualization, and the performance evaluation of the system using an ECG simulator.

4.1 Physical Realization of the Wearable Holter Monitoring System

The proposed smart wearable Holter monitoring system was successfully designed and constructed through the integration of the electronic module, electrode interface, power supply, and wearable enclosure. The final device was compact and light in weight, and it enables continuous monitoring without significantly restricting user movement.

The electronic enclosure securely housed the electronic components while allowing easy access to electrode connections and batteries. The wearable module accommodated the electronic module and the electrodes, with the straps easily adjustable for a secure fit.

Figure 31. The figure shows the completely assembled Printed Circuit Board (Researcher's own construct, 2023).

Figure 34. Front view of the electronics enclosure of the Holter Monitor (Researcher's own construct,

Figure 33. The back view of the electronic enclosure of the Holter Monitor (Researcher's own construct, 2023).

Figure 32. Final design (front and rare view) of the wearable Holter device

4.2 ECG Signal Acquisition Results

The wearable Holter monitoring system successfully acquired ECG signals using the selected electrode configuration. The recorded ECG waveforms exhibited clearly identifiable P waves, QRS complexes, and T waves, which are essential for cardiac rhythm assessment.

The ECG signals showed sufficient amplitude and stability under normal operating conditions, which indicates effective electrode contact and proper functioning of the ECG front-end circuitry. These results confirm that the system is capable of capturing physiologically meaningful ECG signals suitable for continuous monitoring.

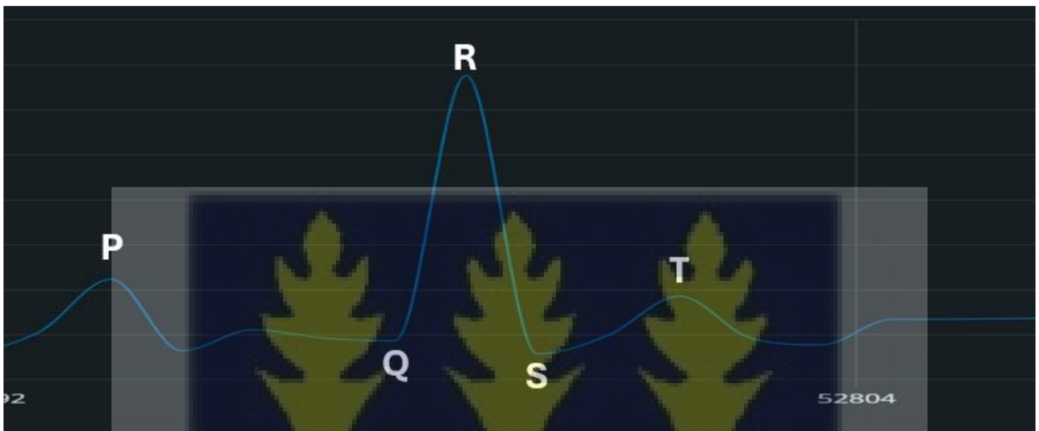


Figure 35. ECG waveform acquired from the wearable Holter monitoring system captured on Arduino running on a laptop

4.3 Wireless Data Transmission and Storage Results

Wireless transmission of ECG and heart rate data from the wearable Holter device to the mobile application was successfully achieved using a Wi-Fi connection and the MQTT communication protocol. Data packets were transmitted in near real-time to enable continuous monitoring through the mobile application.

In addition to wireless streaming, the code embedded in the ESP32 microcontroller was extended to provide the capability of logging data into the attached SD card. Following the acquisition of the ECG signals, the firmware proceeded to append the acquired data in JSON format to the SD card that is integrated into the Holter monitor device. This recorded data included vital information about the heart rate and the heart rhythms recorded during the simulation.

Python script (mentioned in Appendix H) was used as the primary tool for data processing, specifically for managing the JSON-formatted data saved on the SD card. From the raw recorded data, we extracted pertinent information related to heart rate and heart rhythm and generated visual plots in google colab that illustrate these parameters, as seen in Figures 36 and 37.

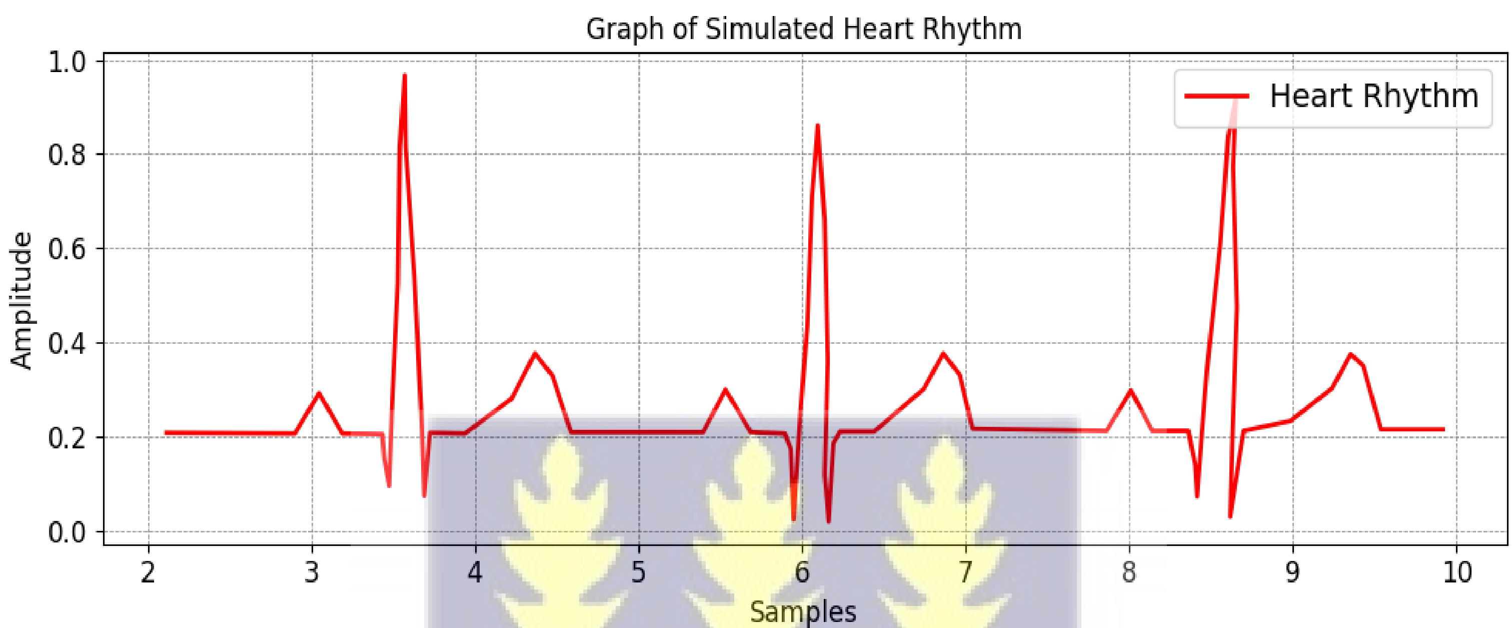


Figure 37. This graph shows the saved simulated heart rhythm data from the SD card

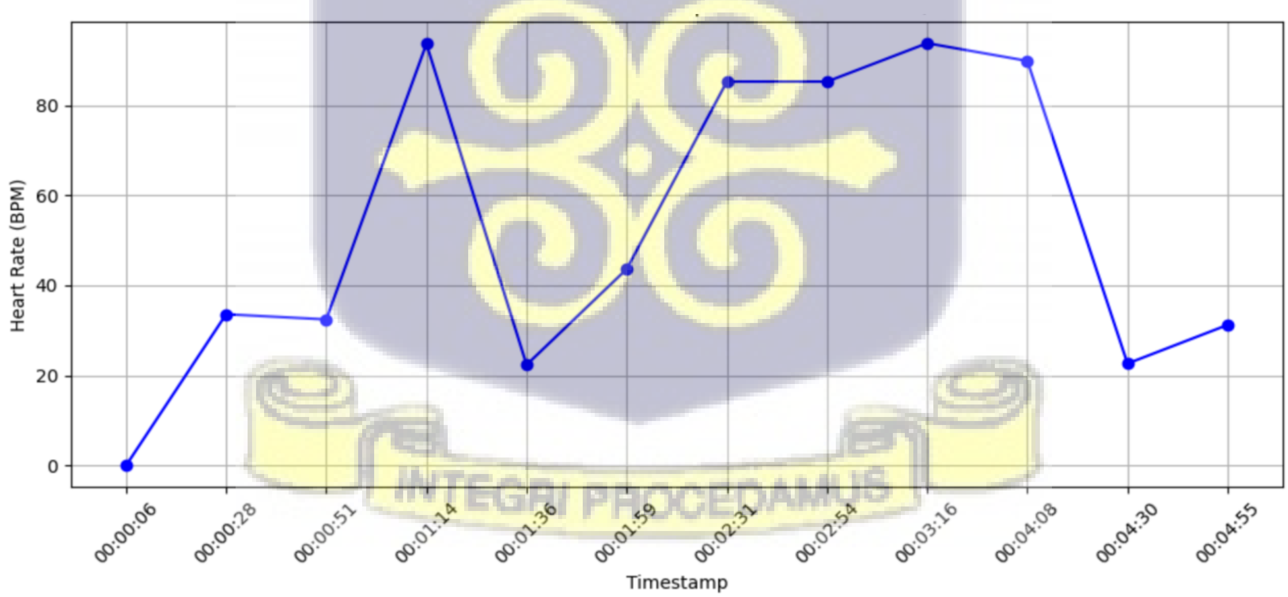


Figure 36.This graph the various heart beats recorded on the SD card.

4.4 Mobile Application Results

The developed Android mobile application served as the primary interface for real-time visualization and interaction with the wearable Holter monitoring system. The application successfully received streamed ECG and heart rate data from the wearable device via the wireless connection.

The main dashboard displayed real-time ECG waveforms and numerical heart rate values. This allows users to monitor cardiac activity continuously. The ECG waveform updated dynamically as new data were received to guarantee accurate representation of the acquired signals.

The mobile application also provided secure user authentication through sign-up and sign-in functionality (see Figure 39). Historical ECG and heart rate data were stored and accessed through the history interface, supporting retrospective analysis without requiring continuous connectivity as shown in Figure 40.

Alert notifications were also generated by the mobile application when heart rate values exceeded or fell below predefined thresholds. These alerts were displayed through the mobile interface, highlighting the role of the smartphone as the primary user interaction and visualization platform.

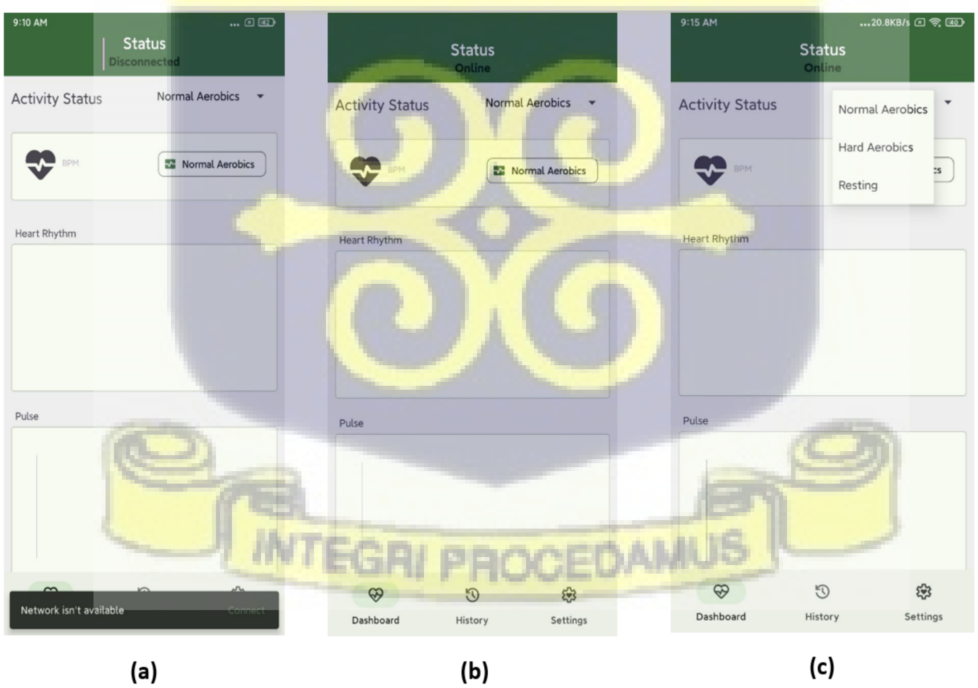


Figure 38. Dashboard of mobile app (a) shows a disconnected status indicating no internet connection. (b) shows an online status indicating the app has access to internet. (c) shows the various activity statuses that can be chosen from.

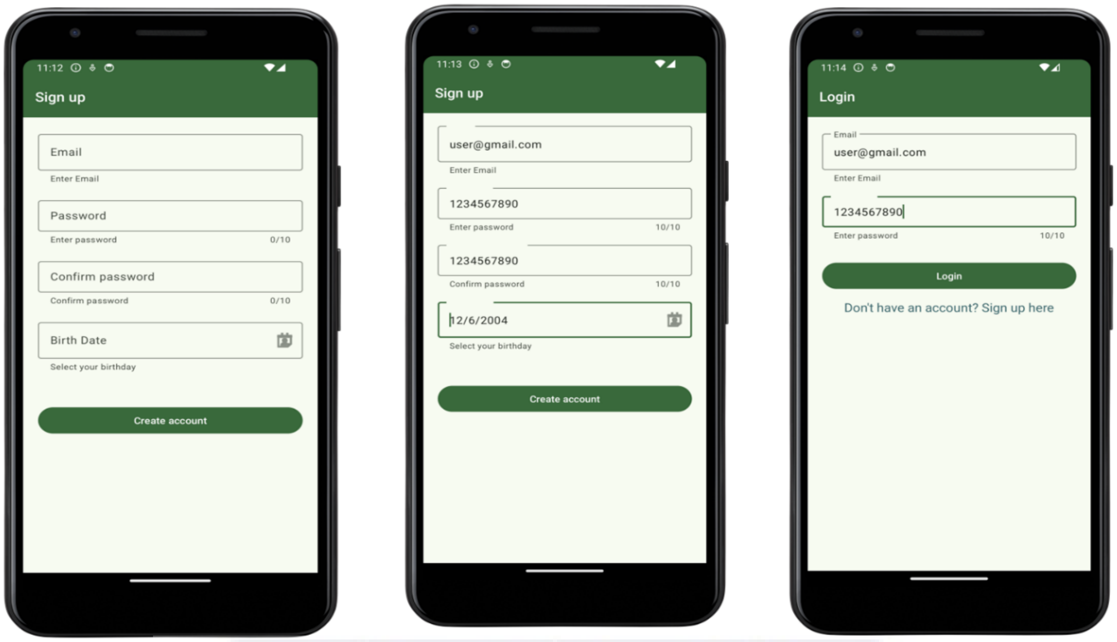


Figure 40. Sign In/ Sign Up face of mobile application showing the necessary information required to set up the application.

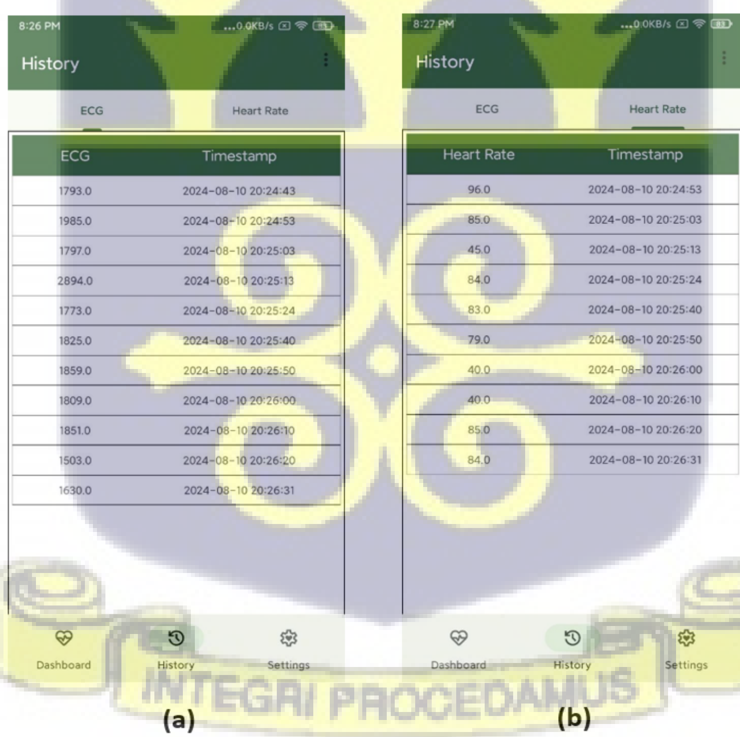


*Figure 39. History tab showing recordings of the **(a) ECG** and **(b) Heart Rate** values.*

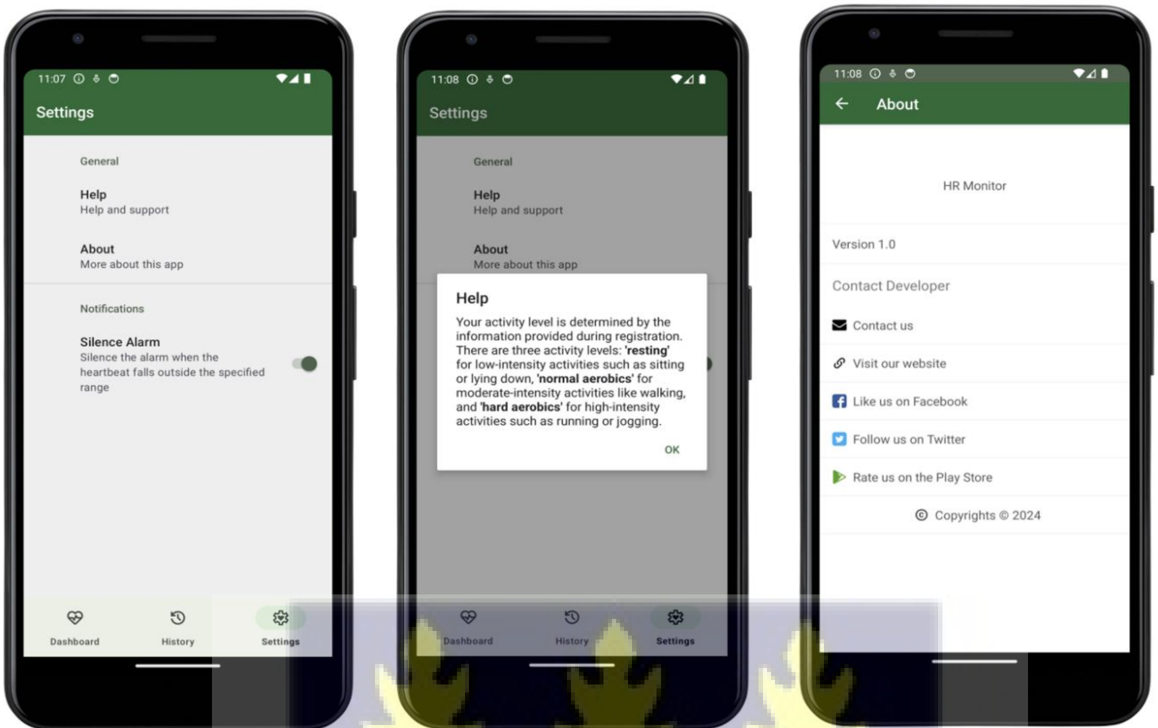


Figure 41. Interface of the Settings of the mobile app where the alarm can be silenced and also shows some information about the mobile app.

4.5 Activity Based Heart Rate Monitoring Results

The activity recognition feature allowed users to select their current activity level, including resting, normal aerobics, and hard aerobics. Based on the selected activity level and the user's age, the system computed appropriate heart rate thresholds using established American Heart Association equations.

During testing, the system adjusted heart rate thresholds according to the selected activity and successfully generated alerts when heart rate values fell outside the computed ranges. This context-aware approach reduced unnecessary alerts during physical activity and improved the relevance of system notifications.

4.6 System Performance Evaluation Using as ECG Simulator

The performance of the proposed wearable Holter monitoring system was evaluated using an ECG simulator, as described in Chapter Three. The simulator generated ECG signals with predefined heart rates and rhythm patterns, allowing objective and repeatable performance assessment.

Figure 42. This figure shows a set-up of the system simulation with the Holter Monitor connected to the laptop via USB cable and the electrodes connected to the corresponding pins on the ECG Simulator

4.6.1 Heart Rate Measurement Accuracy

The simulator was programmed to produce a sequence of heartbeats at a simulated pace of 80 beats per minute, and the recordings from the Holter monitor were analyzed in relation to this established figure. The Holter monitor's measurements of heart rate were documented for 5 minutes at the simulated rate of 80 bpm. The findings indicated a significant level of precision, as the Holter monitor displayed an average heart rate of 79.8 bpm and a mean deviation of ±0.4 bpm. The percentage of accuracy was calculated to be 99.75% using Equation 8.

$$\text{Accuracy (\%)} = \left(1 - \frac{|\text{Measured Value} - \text{Actual Value}|}{\text{Actual Value}}\right) \times 100 \qquad \text{Equation 8}$$

Where, Measured Value = 79.8 bpm

Actual Value = 80 bpm

The average accuracy of 99.75% indicates very close agreement between the Holter monitor readings and the simulated heart rate, demonstrating reliable performance in heart rate detection.

4.6.2 Signal-to-Noise Ratio (SNR)

The signal quality was evaluated by analyzing the signal-to-noise ratio (SNR). To measure the SNR, an artificial sinus rhythm was generated with a particular frequency and amplitude using an ECG simulator, and the resulting data was recorded for 5 minutes. The corresponding graphs on the mobile app were captured and digitized using PlotDigitizer, and the signal and noise components such as baseline wander, muscle noise, or other artifacts were isolated. The power of the clean signal (P_{signal}) and the noise (P_{noise}) was calculated using the mean square values:

$$P_{signal} = \frac{1}{N}\sum_{i=1}^{N} x[i]^2 \qquad \text{Equation 9}$$

$$P_{noise} = \frac{1}{N}\sum_{i=1}^{N} n[i]^2 \qquad \text{Equation 10}$$

Where,

$x[i]^2 = amplitude\ of\ the\ signal\ at\ each\ data\ point$

$(N) = total\ number\ of\ data\ points$

$n[i]^2 = amplitude\ of\ the\ noise\ at\ each\ data\ point$

For a sample data segment, the power values were:

- Signal Power: $P_{signal} = 0.165$
- Noise Power: $P_{noise} = 0.0018$

The SNR was calculated as:

$$SNR(dB) = 10\log_{10}\left(\frac{P_{signal}}{P_{noise}}\right) \approx 19.62\ dB$$

A high signal-to-noise ratio suggests that the quality of the signal is good and it is not easily affected by ambient noise. With an SNR value of approximately 19.62 dB, it is evident that the signal is significantly stronger than the background noise, with the signal being about 92 times more powerful when measured on a linear scale.

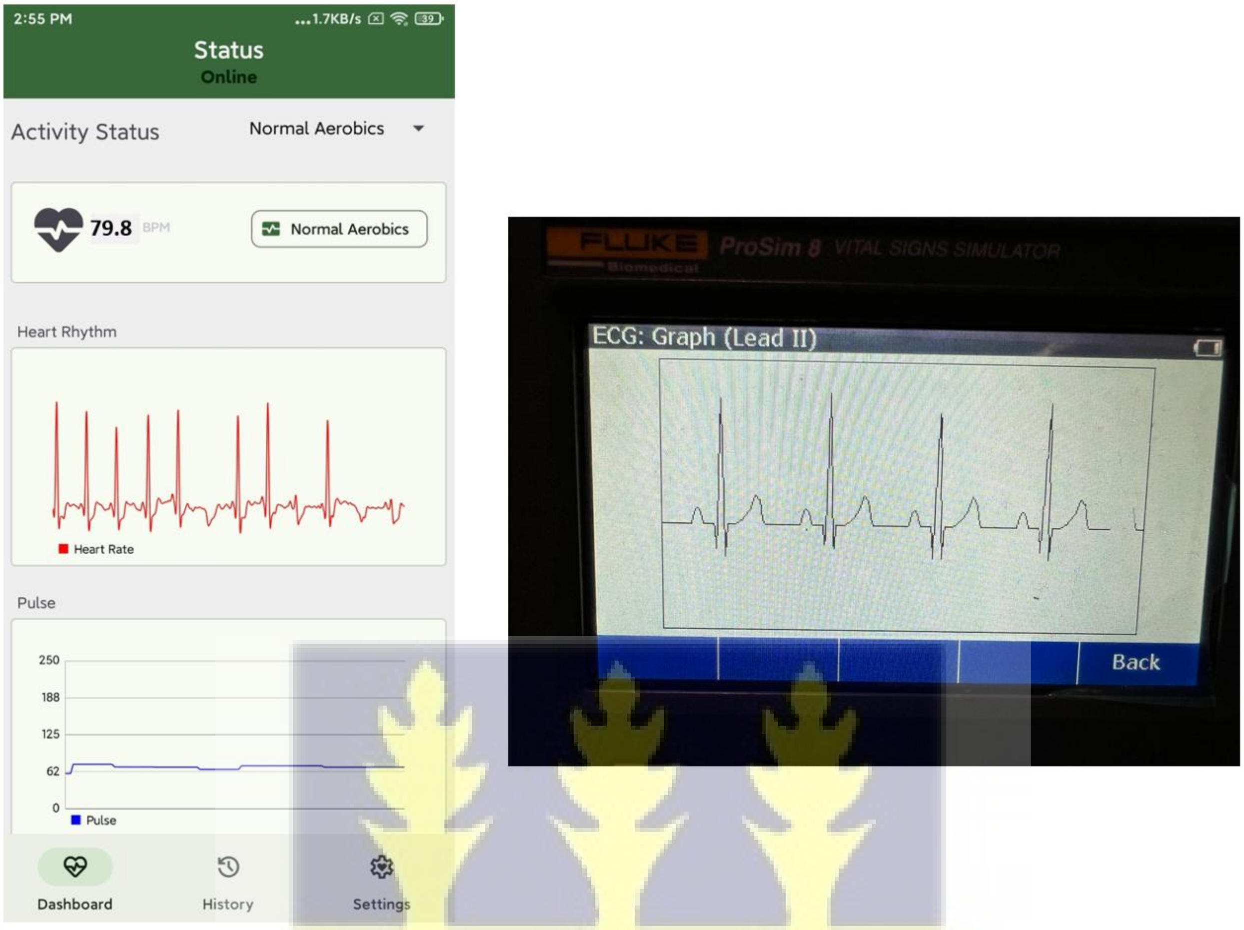
The high SNR suggests that the ECG simulator's signal is of exceptional quality, with minimal interference from noise sources such as baseline wander or muscle artifacts. This ensures that the collected data is reliable and suitable for further analysis, ensuring that any conclusions drawn from this signal are based on accurate and distinct measurements. Essentially, this means that the signal can be easily distinguished from ambient noise, a crucial factor for accurate medical monitoring and diagnosis.

4.6.3 Heart Rhythm Detection

The ECG simulator was once again set up to produce a range of rhythms, such as normal sinus rhythm, atrial fibrillation, ventricular fibrillation, atrial flutter, and ventricular asystole. The graphical results from the Holter monitor were then compared to the reference graphs for these rhythms on the simulator. By examining the ECG patterns on both the simulator's monitor and the mobile application, we can evaluate and guarantee that the signals sent to the Holter monitor are an accurate match to the expected or simulated ECG patterns. This analysis of the displayed graph on the simulator and the output graph from the Holter monitor helps to confirm the accuracy and dependability of the Holter monitor's readings, validating its capability to precisely capture the simulated ECG signals.

The firmware on the ESP32 microcontroller (the source is presented in Appendix G) was designed to initialize the analog-to-digital converter (ADC) to read analog signals present at A0 and convert them to a digital value ranging from 0 to 4095 (since it has a 12-bit ADC).

After starting the setup, the Fluke ECG Simulator produced different heart rhythms and created specific ECG patterns. The Holter monitor recorded these patterns, and the mobile app displayed the transmitted ECG signals in real-time, illustrating the variety of simulated heart rhythms from the Fluke ECG Simulator as seen in Figures 43-45. Monitoring through the mobile app enabled the examination and assessment of the quality and precision of the collected ECG data.



(a) (b)

Figure 43. shows a visual representation of a simulated normal sinus rhythm of 80bpm. This is the most common type of heart rhythm, and it is characterized by a regular heartbeat with a normal rate and rhythm. ***(a)*** *Mobile app dashboard showing the graph at a heartbeat of 79.8 and* ***(b)*** *Fluke ECG simulator display*

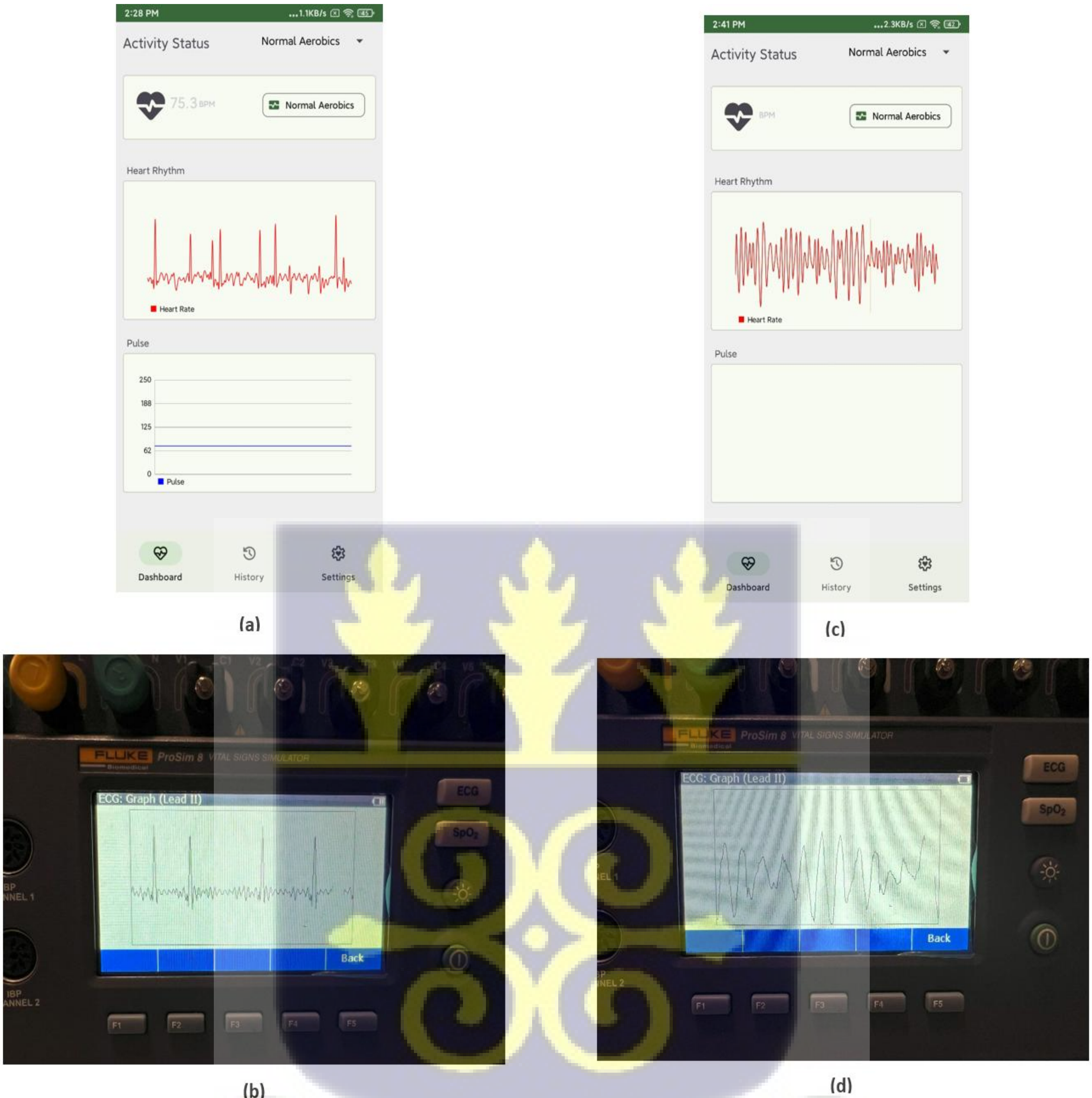


*Figure 44. **(a)** Mobile app **(b)** ECG simulator; shows simulated graphs of Atrial fibrillation which occurs when the atria beat irregularly and out of sync with the ventricles. **(c)** Mobile app **(d)** ECG simulator; shows simulated graphs of Ventricular fibrillation, a serious type of heart arrhythmia that occurs when the ventricles beat erratically and out of sync with each other.*

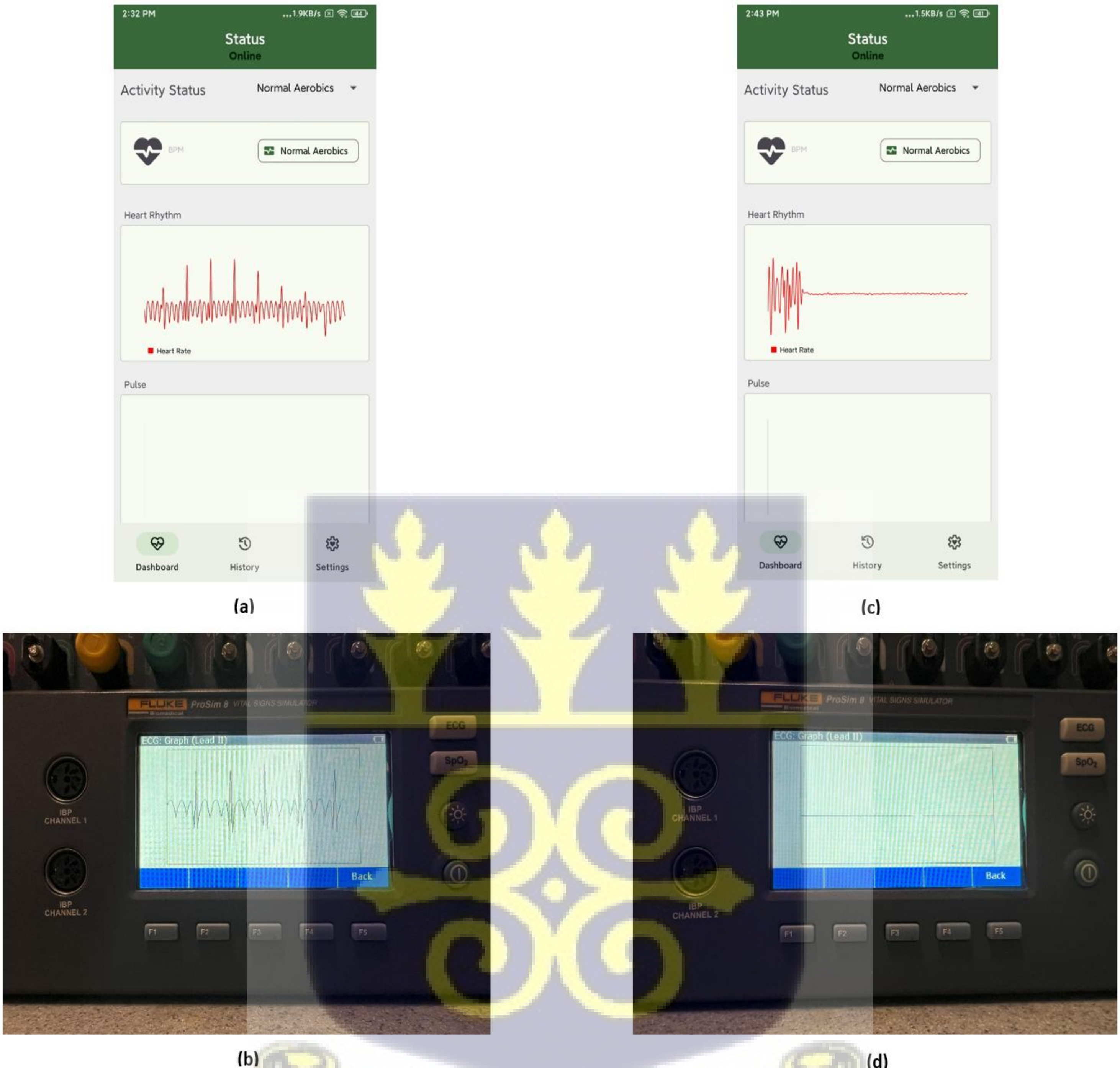


*Figure 45. **(a)** Mobile app **(b)** ECG simulator; shows simulated graphs of Atrial flutter which occurs when the atria beat too fast and irregularly **(c)** Mobile app **(d)** ECG simulator; shows simulated graphs of Ventricular Asystole, a medical condition where there is a complete absence of electrical activity in the ventricles of the heart, resulting in no heartbeat and no cardiac output*

4.6.4 Latency and Responsiveness

The time delay between the ECG simulator generating a heartbeat and the mobile app displaying it was measured using synchronized timestamps. An average latency of 200 milliseconds was determined, which is deemed suitable for real-time monitoring.

4.6.5 Real-Time vs. Offline Analysis in Arrhythmia Detection

Our design offer hybrid functionality which combines real-time monitoring and offline storage. They act to supplement each other, in a paired sense, to get accurate and comprehensive information from the data. Real-time monitoring is achieved through immediate data transmission onto the mobile application using Wi-Fi, and it identifies important events for timely interventions, while the SD card storage gives a complete record for detailed offline analysis.

This combined approach has been shown in literature to be beneficial for both acute care and retrospective evaluation, especially in areas with limited resources where continuous network connection may not be practical. Offline storage resolves potential data continuity difficulties, improving the device's dependability in a variety of scenarios, even though simulated data shows an acceptable delay (200 milliseconds) for real-time streaming.

4.6.6 Continuous Monitoring

The equipment was tested for 24 hours to monitor its continuous performance. It maintained consistent functioning and did not experience any data loss. The battery also remained operational throughout the entire testing period, indicating a reliable battery life for daily usage.

4.6.7 Comparative Analysis of Arrhythmia Detection Methods

The performance and design of the our Holter monitor were compared with existing devices from literature, which was based on the review of prior designs and research efforts in wearable ECG devices. Table 7 shows the comparisons between our Holter monitor and other existing designs found in literature. This comparison focuses on important features like accuracy, data continuity, real-time capabilities, and cost-effectiveness.

Table 7. Comparison of Proposed Holter Monitor with Existing Arrhythmia Detection Methods

Study	Work Done	Data Quality	Real-Time Data Transmission	Data Continuity
Ramli et al. (2011)	Designed a low-cost device using reflectance mode photoplethysmography.	Accuracy affected by finger circumference; PPG-based design prone to noise.	No	No
Mahdy er al. (2018)	Developed a Holter system using an Android app for real-time arrhythmia detection.	Detection accuracy around 78%. Improved with real-time alerts, but still prone to data loss without phone access.	Yes Bluetooth to mobile app	Offline Storage (Mobile Phone)
Noftitasari et al. (2020)	Developed a portable Holter system recording ECG signals for 24 hours.	Provides reliable signal but may miss arrhythmia in future analysis.	Yes Bluetooth to PC	Offline Storage (SD Card Storage)
Demircioglu et al. (2022)	Designed an inexpensive ECG patch with basic threshold-based alerts.	Limited to heart rate detection, difficult to ascertain the type of arrhythmia.	Yes Wi-Fi to Ubidots	Cloud Storage (Ubidots)
Traditional 12-Lead ECG	Not applicable	High-quality, accurate; limited to clinical environments.	Yes, in clinal settings	Limited to brief snapshots
Proposed Holter	Designed a low-cost wearable Holter monitor to aid in early detection of arrhythmia.	Signal processing with AD8232, though further testing is needed.	Yes Wi-Fi to mobile app	Offline Storage (SD Card) and Cloud Storage (MQTTX)

4.7 Discussion

This study aimed to design, implement, and evaluate a low-cost wearable Holter monitoring system capable of continuous cardiac monitoring with real-time mobile visualization to support early detection of arrhythmias. The findings demonstrate that all three project objectives were successfully achieved.

The wearable Holter monitoring device was successfully designed and constructed using low-cost, commercially available components. The integration of the electronic module, electrode interface, power supply, and wearable enclosure resulted in a compact, lightweight, and portable prototype suitable for continuous ambulatory monitoring. Patient comfort, which is critical for long-term compliance in Holter monitoring, was addressed through the wearable design and secure housing of components. The ECG acquisition results confirmed that the selected electrode configuration and AD8232 front-end circuitry were capable of capturing physiologically meaningful ECG signals with clearly identifiable P waves, QRS complexes, and T waves. This confirms that the hardware design meets the fundamental requirements for reliable cardiac signal acquisition.

The mobile application successfully served as the primary platform for ECG visualization and user interaction. Wireless transmission of ECG and heart rate data using Wi-Fi and the MQTT protocol enabled near real-time display of cardiac signals on the mobile dashboard. The inclusion of secure authentication, historical data storage, and activity-based heart rate thresholds improved system usability and reduced false alerts during physical activity. Additionally, the hybrid data handling approach, which combines real-time mobile visualization with local SD card storage, ensured data continuity during network interruptions, making the system suitable for deployment in low-resource settings with unreliable internet connectivity.

Performance evaluation using an ECG simulator demonstrated strong system reliability. The heart rate measurement accuracy of 99.75%, with a deviation of only ±0.4 bpm from the simulated reference value, indicates high precision suitable for continuous cardiac monitoring. Signal quality analysis revealed a signal-to-noise ratio of approximately 19.65 dB, confirming effective noise suppression and reliable ECG waveform interpretation. The system accurately reproduced multiple rhythm types, including normal sinus rhythm, atrial fibrillation, ventricular fibrillation, atrial flutter, and ventricular asystole, supporting its potential for early arrhythmia detection.

Furthermore, the average system latency of approximately 200 milliseconds falls within acceptable limits for real-time cardiac monitoring and does not compromise timely alert generation.

Overall, the results demonstrate that the proposed wearable Holter monitoring system offers a practical, low-cost solution for continuous cardiac monitoring. By combining reliable ECG acquisition, real-time mobile visualization, and offline data storage, the system addresses key limitations of existing low-cost monitoring solutions.

4.8 Project Evaluation

The major objectives for the project (Summarized in Figure 9) were categorized into sections: safety, cost-effectiveness, and performance. This section (presented in Table 8) evaluates the project outcomes by comparing them with stated project objectives, using the specified requirements and specifications.

Table 8. Evaluation of Project Outcomes

Objective	Specification		Actual	Achieved/ Comments
Performance	Lightweight	Weight ≤100g	Prototype weight approximately 100g	Yes
	Portable	≤100mm x 60mm x80mm	64mm x 31mm x 74mm	Yes
	Measures heart rhythm and beats per minute	-	Functionality achieved	Yes
	Number of Electrodes	≤3	ECG sensor with 3 electrodes	Yes
	Battery Life	≤1000mAh	690mAh	No. The choice of a 9V Energizer battery resulted in a lower overall capacity
	Easy to use	Start/Stop options	Prototype constructed with a general switch	Yes
	Data Storage	≥4GB	32GB	Yes
	AD Conversion	≥10bits	ESP32 with ADC of 1 2 bits	Yes
	Processing Speed	>100MHz	ESP32 with a clock speed of 160MHz	Yes
	RAM size	>128Kb	ESP32 with a RAM size of 520Kb	Yes
	Wireless Connectivity	Bluetooth 4.0 Wi-Fi 802.11n	Wi-Fi 802.11n	No. integrating both Bluetooth and Wi-Fi on ESP32 increased the size of the source code.
Safety	Fail safe	Button	Prototype constructed with a general switch	Yes
	Safe to User	Weight≤100g	Prototype weight approximately 100g	Yes
	Stop Mid Operation	Button	Prototype constructed with a general switch	Yes
Cost-Effectiveness	Affordable	≤500GHS	Prototype construction cost GHS 468	Yes

4.9 Prototype Construction Cost

The prototype was constructed to test the functionality of the chosen design concept and the cost incurred is presented in Table 9.

Table 9. Bill of materials

No.	Part	Quantity	Description	Cost
1	OLED	1	Displays the output	17
2	ESP32	1	Programmable microprocessor	19
3	AD8232 sensor	1	Measures heart contractions	85
4	SD Card Module	1	Interfaces SD card with microprocessor	10
5	SD Card	1	Stores data	30
6	Buzzer	1	Sounds alert	10
7	LED	2	Indicates status or specific conditions	8
8	Switch	1	Controls device ON/OFF	5
9	Alkaline Battery	1	Power source	20
10	PCB	1	Circuit board	150
11	Capacitors	7	Store and release electrical energy	12
12	Resistors	7	Regulates current flow	12
13	REG1117	1	Provides 3.3V from alkaline battery	5
14	MC78M05ABDTG	1	Provides 5V from the alkaline battery	5
15	Wearable Pouch	1	Houses the device	80
TOTAL				**468**

At the end of the design project, GHC 468.00 was spent on the building of the Holter monitor.

4.10 Project Work Cost

Some requirements of the project were readily available while some were paid for. The total project cost was GHS 868.00. Table 21 (Refer to Appendix F) provides further details.

CHAPTER FIVE

CONCLUSION AND RECOMMENDATION

5.1 Outcome

After identifying the problem and drafting customer requirements, the engineering design process was successfully applied to solve the identified problem. A prototype of the selected concept was successfully developed to evaluate the primary functionality of the chosen concept. Consequently, the primary objective of the project, which involved developing a hardware for monitoring and storing cardiac rhythm and heart rate, was effectively accomplished.

5.2 Challenges

The difficulties faced included learning and perfecting a new set of skills, particularly in circuit board design and the use of Android Studio. Accessing an ECG simulator also proved to be quite challenging. Additionally, unanticipated delays and impediments were encountered in the process of acquiring the essential components for project completion. These unforeseen obstacles added a significant level of complexity to the project and thus called for an increased amount of time and resources to ensure its successful completion. However, despite these obstacles, we remained steadfast and worked tirelessly to surmount these challenges.

5.3 Conclusion

The project saw the successful creation and integration of a cost-efficient wearable Holter Monitoring System for detecting arrhythmia at an early stage. The design allows for the capturing, storing, and transmitting of signals, giving users the ability to monitor their ECG data and track changes in heart rate using a smartphone app. In contrast to traditional devices, this system is designed with a focus on being lightweight and user-friendly, prioritizing comfort and long-term wearability.

The implementation of a three-lead electrode setup decreases disruptions, while including a 100 nF decoupling capacitor enhances signal filtration from the AD8232 sensor. The wireless transmission enhances the adaptability and functionality of the Holter Monitor. Additionally, the device can serve as an alert system for everyday activities or physical exertion, adding another noteworthy attribute. Its multi-functional capabilities broaden its practicality beyond that of standard Holter monitors.

In summary, this wearable Holter Monitoring system shows potential progress in the early detection of arrhythmias, blending technical accuracy with user-friendly features and increased capabilities.

5.4 Recommendations

In future work, an important addition would be the use of machine learning algorithms to categorize various forms of arrhythmias. This approach will not just aid the system in identifying abnormal heart rhythms, but also in organizing and classifying them, leading to a greater comprehension of cardiac disorders.

In addition, it will be essential to carry out clinical studies with human subjects to evaluate the device's effectiveness in real-life situations. These studies will offer important information on its precision and dependability in identifying irregular heart rhythms in a variety of patient groups. It is imperative to prioritize ethical considerations and patient well-being throughout these studies. Approval from institutional review boards (IRBs) and informed consent from participants will be required.

Concentrating on these specific areas for future enhancements will significantly improve the performance of the wearable Holter Monitoring system in early arrhythmia detection and enhance its efficacy in clinical environments.

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APPENDIX

APPENDIX A: SPECIFICATION TABLE FOR HOLTER MONITOR

Table 10. Specification Table for Holter Monitor System

Requirement	Origin	Importance	Unit	Target Value	Demand/ Wish
Input Unit					
Non-Invasive electrodes	User	9	-	-	Demand
Number of electrodes	Designer	5	Integer	≤3	Wish
Electrode contact pressure	Designer	8	mmHg	50 - 150	Demand
Electrical safety	User	9	-	-	Demand
Electrode size	Designer	9	mm	≤15	Demand
Energy					
Voltage supply	Designer	5	V	≤12	Demand
Battery capacity	User	8	mAh	≤1000	Demand
Power consumption	Designer	8	mW	50 - 150	Demand
Battery life	User	7	Hours	≥48	Wish
Signal Processing Unit					
Heart Rate Recording	User	9	BPM	-	Wish
Bit resolution	Designer	9	bits	≥10	Demand
Noise reduction	Designer	8	Qualitative	Analog filtering	Wish
Wireless connectivity	User	8	Qualitative	Bluetooth 4.0 WiFi 802.11n	Demand
AD Conversion	Designer	5	bits	≥10	Demand
RAM size	Designer	6	Kb	>128	Demand
Processing Speed	Designer	6	MHz	>100	Wish
Whole System					
Number of parts	Designer	2	-	≤5	Wish
Assembly Steps	Designer	2	-	≤4	Wish
Size	User/ Designer	7	mm	≤100x60x15	Wish
Weight	User/ Designer	6	g	≤100	Wish
Data storage	User	7	GB	≥4	Wish
Material Cost	Designer	6	GHS	≤500	Wish
Ease of use	User	7	-	-	Wish
Operating temperature	Designer	7	°C	15 - 40	Demand
Storage temperature	Designer	5	°C	0 - 50	Wish

APPENDIX B: HOUSE OF QUALITY FOR HOLTER MONITOR

Table 11. House of Quality for Holter Monitor

KEY
High Correlation - 9
Moderate Correlation - 4
Low Correlation - 1

Customer Requirements (WHAT)	Engineering Parameters (HOW)	Relationship Strength
Detect heart rhythm	Type of sensor	9
	Number of leads	4
Non-invasive design	Attachment mechanism	9
	Electrode material	9
Secure and dependable electrode contact	Electrode material	9
	Adhesive strength	9
Electrical Safety	Energy Source	1
	Insulation	9
	Shape and Sharp edges	4
Discrete	Size	4
	Weight	4
Affordable	Material cost	9
	Manufacturing cost	9
	Component Selection	1
Comfortable	Material	4
	Size	4
	Weight	4
Self-Operational	Automation	9
Easy to Maintain	Accessibility of components	4
	Software design	9
Supports Wireless Connectivity	Wireless communication protocol	9
	Data transmission rate	4
Long Battery Life	Battery capacity	9
Data Storage Capacity	Storage technology	4
	Memory	4
Easy to Operate	Automation	4
	User interface	9
	Steps in operation	1

APPENDIX C: CONCEPT SELECTION USING DECISION MATRICES

Table 12. Table evaluating options for the Type of Electrode

SUB FUNCTIONS		OPTION 1	OPTION 2	OPTION 3
TYPE OF ELECTRODE		Gold Electrodes	Ag/AgCl Electrodes	Graphite Electrodes
Criteria	**Weight (1-5)**			
Signal Quality	5	5 x 5 = 25	5 x 5 = 25	5 x 5 = 25
Skin Irritation	4	4 x 4 = 16	4 x 4 = 16	2 x 4 = 8
Convenience	3	2 x 3 = 6	4 x 3 = 12	4 x 3 = 12
Longevity	3	2 x 3 = 6	4 x 3 = 12	2 x 3 = 6
Cost	2	2 x 2 = 4	4 x 2 = 8	4 x 2 = 8
Compatibility	4	5 x 4 = 20	5 x 4 = 20	3 x 4 = 12
Ease of Use	4	2 x 4 = 8	4 x 4 = 16	4 x 4 = 16
Recommended Use	5	2 x 5 = 10	5 x 5 = 25	4 x 5 = 20
Total Score		**95**	**134**	**107**

ECG electrodes play a pivotal role in capturing accurate heart signals. The distinctive characteristics of various electrode types, such as Ag/AgCl, gold, and graphite, affect their usefulness for ECG collection.

Ag/AgCl electrodes are the best choice for ambulatory monitoring because their electrode potentials are consistent and predictable, they are biocompatible, they don't irritate the skin, they have high signal quality, they respond quickly, and they are cost-effective, especially when compared to other disposable electrodes(Elango et al., 2023; Lee et al., 2022). Although gold electrodes possess chemical inertness and biocompatibility, they are rarely used in electrocardiogram (ECG) collection due to their higher cost and complexity(Elango et al., 2023). Graphite electrodes have shown potential for improved ECG signal quality and lower contact impedance, but their use may be limited due to their relatively new development and potential manufacturing challenges(Celik et al., 2016).

Hence, the chosen option for the subfunction type of electrode was Ag/AgCl.

Table 13. Table evaluating options for the Number of Electrodes

SUB FUNCTIONS		OPTION 1	OPTION 2	OPTION 3	OPTION 4
NUMBER OF ELECTRODES		Twelve-lead Configuration	Five-lead Configuration	Three-lead Configuration	Single-lead Configuration
Criteria	**Weight (1-5)**				
Signal Detail	5	5 x 5 = 25	5 x 5 = 25	4 x 5 = 20	2 x 5 = 10
Diagnostic Capability	4	5 x 4 = 20	5 x 4 = 20	4 x 4 = 16	2 x 4 = 8
User Comfort	3	0 x 3 = 0	2 x 3 = 6	4 x 3 = 12	5 x 3 = 15
Ease of Use	3	2 x 3 = 6	2 x 3 = 6	4 x 3 = 12	5 x 3 = 15
Total Score		**51**	**57**	**60**	**48**

The selection of the electrocardiogram (ECG) lead configuration in a Holter monitor has a substantial impact on the extent of cardiac data collected and the level of comfort experienced by the patient throughout the monitoring process.

In this project, we evaluated the single-lead, 3-lead, 5-lead, and 12-lead configurations. We compared the advantages and disadvantages of each configuration to determine the most suitable option. The single-lead configuration is the simplest and most comfortable for the patient, but it provides limited information about the heart's electrical activity. The 5-lead configuration provides more detailed information and is often used in specific clinical situations. The 12-lead configuration offers the most comprehensive cardiac information but may be less comfortable for the patient due to the larger number of electrodes. After careful consideration, we chose the 3-lead ECG configuration(Francis, 2016). It offers a balance between comfort and information, making it suitable for routine monitoring. This choice ensures sufficient cardiac data collection while maintaining patient comfort, serving as the most practical and balanced option for our Holter monitor project.

Table 14. Table evaluating options for the Power Source

SUB FUNCTIONS		OPTION 1	OPTION 2	OPTION 3
POWER SOURCE		Rechargeable Batteries	Solar	Disposable Batteries
Criteria	**Weight (1-5)**			
Reliability	5	4 x 5 = 20	2 x 5 = 10	5 x 5 = 25
Safety	5	4 x 5 = 20	4 x 5 = 20	5 x 5 = 25
Affordability	4	2 x 4 = 8	4 x 4 = 16	5 x 4 = 20
Longevity	4	4 x 5 = 20	4 x 4 = 16	4 x 4 = 16
Convenience	4	4 x 4 = 16	4 x 4 = 16	2 x 4 = 8
Environmental Impact	3	4 x 3 = 12	5 x 3 = 15	2 x 3 = 6
Total Score		**96**	**93**	**100**

The selection of batteries for a Holter monitor is a critical factor in maintaining continuous and dependable monitoring of an individual's cardiac activity. In this project, we considered the options of rechargeable batteries, solar batteries, and disposable batteries for the Holter monitor.

After careful analysis, we concluded that disposable batteries were the best choice for the Holter monitor. While rechargeable batteries may seem like a more sustainable option, they require frequent charging and may not hold a charge for an extended period of time. Solar batteries, on the other hand, rely on sunlight and may not be suitable for all environments or situations. Disposable batteries, although less eco-friendly, provide a consistent and reliable power source for the monitor, ensuring continuous monitoring without the need for frequent replacements or recharging. In addition, their portability, being lightweight and compact, allows for wearable designs without adding extra size or weight, unlike solar or rechargeable alternatives. Again, they are readily available and easy to replace in case of emergencies. Finally, their safety measures decrease the likelihood of overheating or electrical burns in the event of damage during usage, as opposed to certain types of rechargeable batteries.

Table 15. Table evaluating options for the Fabric

SUB FUNCTIONS		OPTION 1	OPTION 2	OPTION 3
FABRIC		Cotton	Spandex	Nylon
Criteria	**Weight (1-5)**			
Comfort	5	5 x 5 = 20	4 x 5 = 20	4 x 5 = 20
Durability	4	4 x 4 = 16	3 x 4 = 12	5 x 4 = 20
Breathability	4	5 x 4 = 20	4 x 4 = 16	2 x 4 = 8
Skin-Friendly	3	5 x 4 = 20	4 x 3 = 12	4 x 3 = 12
Total Score		**76**	**60**	**60**

Cotton, spandex, and nylon are all materials with different properties that can affect the comfort, breathability, and functionality of a wearable Holter monitor.

Cotton is notable for its comfort, breathability, and gentle feel on the skin, making it an excellent choice for a wearable Holter monitor. Because of its great moisture absorption, smooth texture, and skin-friendliness, it is beneficial for extended use, lowering the chance of discomfort and promoting regular use for continuous data gathering. Although spandex is known for its exceptional stretch and ability to conform to the body, and nylon is recognized for its durability and moisture-wicking capabilities, the overall comfort and user-friendly nature of cotton, particularly in areas that come into direct contact with the skin, surpass the specific benefits of spandex and nylon in the context of designing a wearable for a Holter monitor(Vijayan et al., 2021). This ensures both comfort and accuracy in data collection.

Table 16. Table evaluating options for the Microcontroller

SUB FUNCTIONS		OPTION 1	OPTION 2	OPTION 3
MICROCONTROLLER		Arduino Uno	ESP32	Raspberry Pi
Criteria	**Weight (1-5)**			
Processing Power	5	2 x 5 = 10	4 x 5 = 15	5 x 5 = 25
Low Power Consumption	5	5 x 5 = 25	4 x 5 = 15	2 x 5 = 10
Connectivity	4	0 x 4 = 0	5 x 4 = 20	5 x 4 = 20
Ease of Programming	4	5 x 4 = 20	5 x 4 = 20	4 x 4 = 16
Cost	3	5 x 3 = 15	4 x 3 = 12	2 x 3 = 6
Total Score		**70**	**82**	**77**

When choosing a platform for designing a Holter monitor, considerations of power consumption, computational capability, sensor integration, and cost are crucial factors. In this study, a comparative analysis was conducted on three widely used Node MCUs, namely the ESP32, Arduino Uno, and Raspberry Pi. The selection of the ESP32 was based on its advantageous mix of features, including its efficient power usage, integrated wireless communication capabilities, and robust processing capabilities inside a compact physical design. Additionally, its extensive support ecosystem made it the ideal choice for developing a user-friendly next-generation Holter monitor platform for medical applications(Systems, 2016). The ESP32's capacity to effectively manage the computing demands of the monitor, seamlessly integrate diverse sensors, and its cost-effectiveness have firmly established its status as the choice platform for this project.

Table 17. Table evaluating options for the Casing

SUB FUNCTIONS		OPTION 1	OPTION 2	OPTION 3
CASING		Polylactic Acid	Fabric	Plastic
Criteria	**Weight (1-5)**			
Durability	5	4 x 5 = 20	2 x 5 = 10	4 x 5 = 20
Water Resistance	4	4 x 4 = 16	2 x 4 = 8	4 x 4 = 16
Customizability	4	2 x 4 = 8	4 x 4 = 16	2 x 4 = 8
Environmental Impact	4	5 x 4 = 20	3 x 4 = 12	3 x 4 = 12
Comfort	3	2 x 3 = 6	4 x 3 = 12	2 x 3 = 6
Total Score		**70**	**58**	**62**

Among the materials we looked at for the electronic module case of our Holter monitor, polylactic acid (PLA) proved to be the best choice, beating out fabric and plastic. This is mostly because of its unique properties, though it does have some built-in limitations. The use of polylactic acid (PLA) as the primary material for the electronic enclosure case has several benefits, including its biodegradability, ability to be customized through 3D printing, lightweight properties, and biocompatibility. Although PLA may present challenges related to durability and environmental decomposition, it is in line with sustainability objectives that prioritize user convenience in the design of an enclosure for the electronic module(Joseph et al., 2023).

Table 18. Table evaluating options for the Alert

SUB FUNCTIONS		OPTION 1	OPTION 2	OPTION 3
ALERT		Audio	Visual	Audio + Visual
Criteria	**Weight (1-5)**			
Notification Clarity	5	4 x 5 = 20	4 x 5 = 20	5 x 5 = 25
User Disruption	4	2 x 4 = 8	4 x 4 = 16	4 x 4 = 16
Customization	3	2 x 3 = 6	2 x 3 = 6	4 x 3 = 12
Total Score		34	42	53

Holter monitors use different alerts to notify wearers or healthcare providers of irregularities in heart activity. Audio alerts, visual alerts, and audio-visual alerts were considered in this design. Audio-visual alerts are the preferred design choice for Holter monitors, combining auditory signals like beeps or tones with on-screen visual cues such as flashing icons or color changes. The dual-alert system guarantees a thorough and prompt alerting of abnormalities in cardiac activity, offering both audio and visual input simultaneously, irrespective of the wearer's active engagement with the monitor. The combination of auditory and visual alarms provides a strong and efficient method of rapidly notifying both the individual using the device and healthcare professionals. This enables timely action or attention to be given to any identified abnormal cardiac rhythms(Francis, 2016).

Table 19. Table evaluating options for the Storage

SUB FUNCTIONS		OPTION 1	OPTION 2	OPTION 3
DATA STORAGE		Solid State Drive	Mini SD Card	Micro SD Card
Criteria	**Weight (1-5)**			
Capacity	5	5 x 5 = 25	4 x 4 = 16	2 x 5 = 10
Accessibility	4	4 x 4 = 16	5 x 4 = 20	4 x 4 = 16
Reliability	4	4 x 4 = 16	4 x 4 = 16	2 x 4 = 8
Portability	3	4 x 3 = 12	5 x 3 = 15	5 x 3 = 15
Affordability	3	2 x 3 = 6	4 x 3 = 12	4 x 3 = 12
Total Score		**75**	**79**	**61**

When designing a Holter monitor, the choice between Solid State Drives, mini-SD cards, and micro-SD cards is based on things like cost, performance, durability, storage space, and physical size. The use of a mini-SD card within the architecture of a Holter monitor presents a compact, cost-effective, and energy-efficient approach while simultaneously offering an adequate level of storage capacity for uninterrupted monitoring. The compact dimensions, cost-effectiveness, and lower energy usage of this device render it a very suitable option for space-limited devices such as Holter monitors, while still maintaining a considerable storage capacity.

Table 20. Table evaluating options for the Display Unit

SUB FUNCTIONS		OPTION 1	OPTION 2	OPTION 3
DISPLAY UNIT		OLED	Smartphone App	OLED + Smartphone App
Criteria	**Weight (1-5)**			
Functionality	5	4 x 5 = 20	5 x 5 =20	4 x 5 = 20
Readability	4	4 x 4 = 16	4 x 4 = 16	5 x 4 = 20
Interactivity	4	2 x 4 = 8	5 x 4 = 20	5 x 4 = 20
Power Efficiency	4	4 x 4 = 16	4 x 4 = 16	2 x 4 = 8
User Interface	4	2 x 4 = 8	5 x 4 = 20	4 x 4 = 16
Visibility	3	2 x 3 = 6	4 x 3 = 12	4 x 3 = 15
Aesthetics	3	2 x 3 = 6	4 x 3 = 12	4 x 3 = 15
Convenience	3	2 x 3 = 6	4 x 3 = 12	4 x 3 = 15
Total Score		**86**	**128**	**129**

The use of both OLED technology and a smartphone application in the design of a Holter monitor effectively leverages the unique capabilities of each component, therefore augmenting the entire monitoring experience. The OLED display provides instantaneous data visualization on the device itself, enabling rapid access to crucial information such as heart rate, eliminating the need for an additional device. This feature is advantageous for those who have a preference for immediate response and for people who find it cumbersome to access a smartphone in certain situations. In contrast, the smartphone application offers a more expansive screen, facilitating in-depth data analysis, extensive archival of past information, and user-centric interfaces that enhance navigability and understanding. The application enables the ongoing observation of health indicators over an extended period, as well as the exchange of data with healthcare professionals. Additionally, it fosters a deeper comprehension of the collected data, which is crucial for complete health maintenance. The integration of both functionalities enables users to promptly access essential data on their device, while also providing the choice for more extensive analysis and storage on their smartphones(Mahdy et al., 2018). This accommodates a wider array of user preferences and requirements in the monitoring of their heart health.

APPENDIX D: SCHEMATIC DIAGRAM OF HOLTER MONITOR

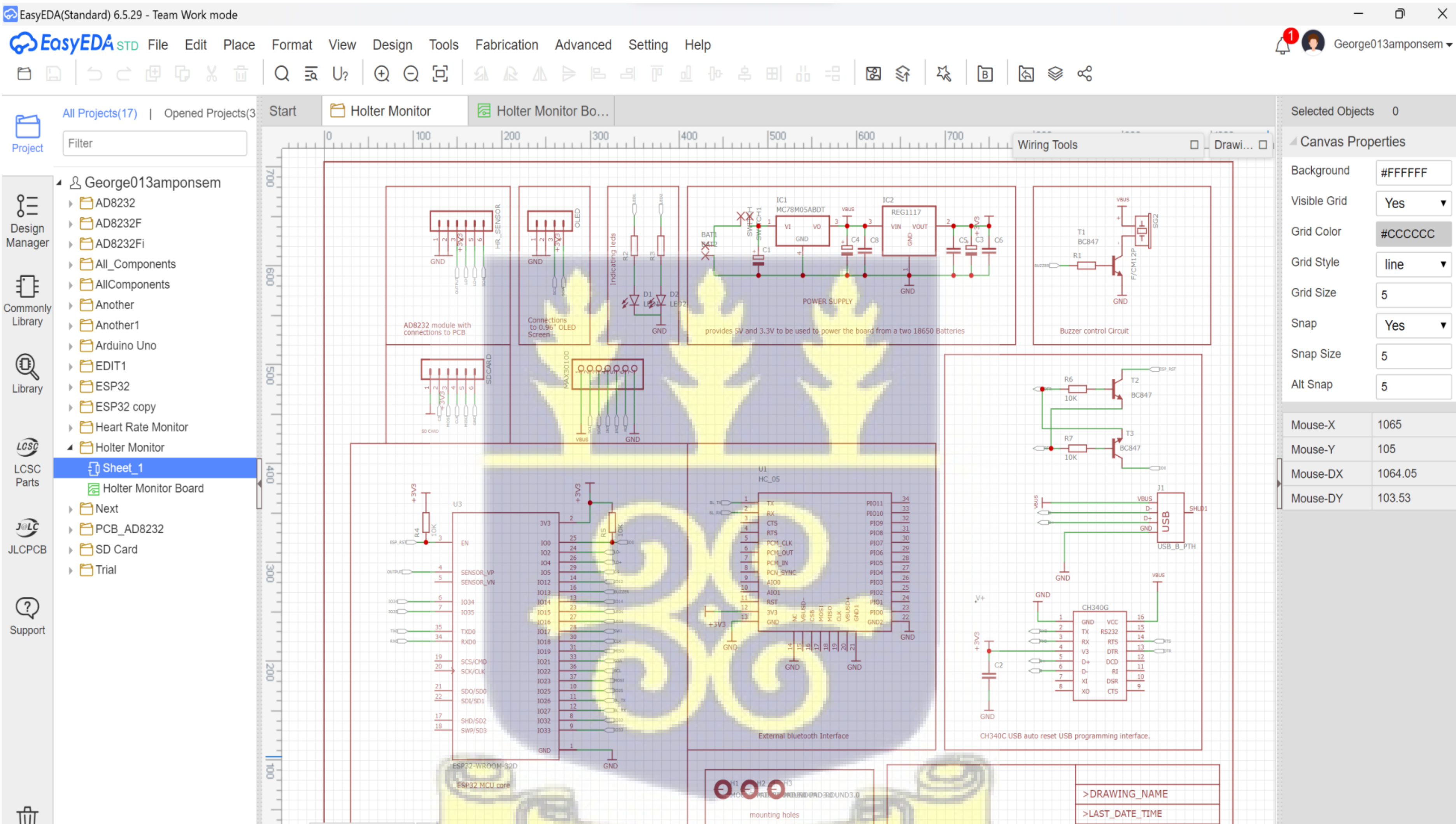


Figure 46. Schematic Diagram of Holter Monitor

APPENDIX E: LAYOUT OF THE ENCLOSURE

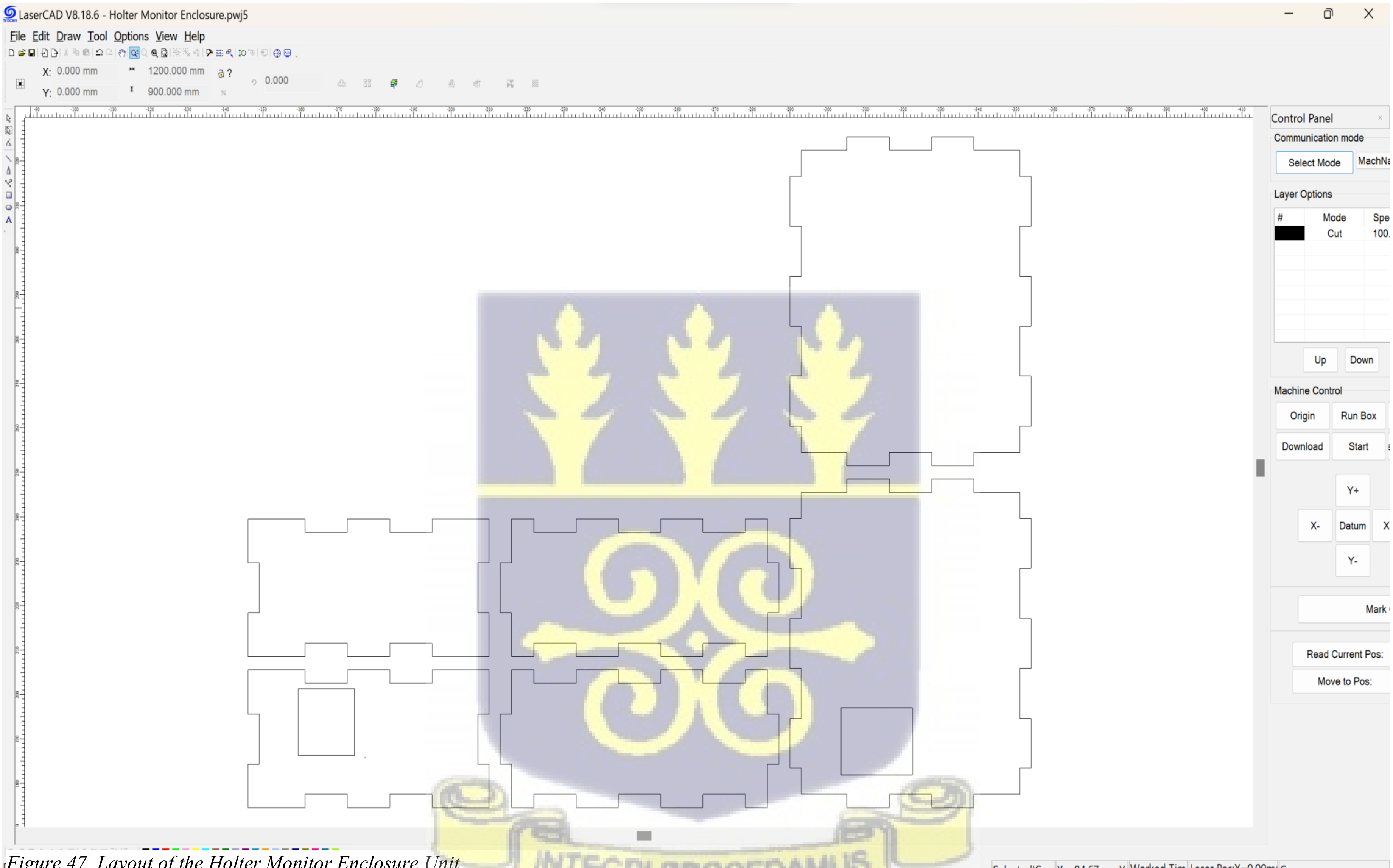


Figure 47. Layout of the Holter Monitor Enclosure Unit

APPENDIX F: PROJECT WORK COST

Table 21. Shows the entire cost incurred during the project

No.	Requirement	Function	Quantity	Cost (GHS)
Research				
	Internet Service	Research Information sources	-	200
	Laptop	Program installation and use	1	Available
Detail Design				
	Arduino (Version 2.1.0)	MCU Programming	1	Available
	EasyEDA (Version 6.5.29)	PCB Design	1	Available
	Android Studio	Mobile App Design	1	Available
	Design Materials	Prototype construction	-	468
Presentation and Reports				
	Microsoft Office 2023	Writing reports and presentations	1	Available
	Paper	Printing reports	-	200
TOTAL				**868**

APPENDIX G: SOURCE CODE FOR FIRMWARE ON HOLTER MONITOR

```
#include <PubSubClient.h>
#include <WiFi.h>
#include <Wire.h>
#include <Adafruit_SSD1306.h>
#include "FS.h"
#include "SD.h"
#include "SPI.h"

// Callback methods prototypes
void t1Callback();
void mqttthings();
void freqDetec();
void normalizeData();

#define SCREEN_WIDTH 128 // Display width, in pixels
#define SCREEN_HEIGHT 32 // Display height, in pixels
#define OLED_RESET -1     // Reset pin # (or -1 if sharing Arduino reset pin)
#define SCREEN_ADDRESS 0x3C // OLED I2C address

Adafruit_SSD1306 display(SCREEN_WIDTH, SCREEN_HEIGHT, &Wire, OLED_RESET);

#define REPORTING_PERIOD_MS 1000
#define WIFISSID "Galaxy"
#define PASSWORD "helloworld"
#define DEVICE_LABEL "esp32"
#define VARIABLE_LABEL_1 "heart_rhythm"
#define VARIABLE_LABEL_2 "pulse"
#define MQTT_CLIENT_NAME "EI_OXMO"
#define SENSOR A0
```

```
int led1 = 15;
int led2 = 16;
int buzzer = 13;
char mqttBroker[] = "broker.hivemq.com";
char payload[700];
char topic[150];

WiFiClient wifiClient;
PubSubClient client(wifiClient);

#define MAX_BUFFER 100

uint32_t prevData[MAX_BUFFER];
uint32_t sumData = 0;
uint32_t maxData = 0;
uint32_t avgData = 0;
uint32_t roundrobin = 0;
uint32_t countData = 0;
uint32_t period = 0;
uint32_t lastperiod = 0;
uint32_t millistimer = millis();
double frequency;
double beatspermin = 0;
uint32_t newData;
int ledState = LOW;

unsigned long previousMillis = 0;
const long interval = 10000;
uint32_t tsLastReport = 0;

// Space to store values to send
```

```
char str_val_1[10];
char str_val_2[10];
int flag = 0;

void callback(char* topic, byte* payload, unsigned int length) {
  Serial.print("Message arrived [");
  Serial.print(topic);
  Serial.print("] ");
  for (int i = 0; i < length; i++) {
    Serial.print((char)payload[i]);
  }
  Serial.println();
}

void reconnect() {
  while (!client.connected()) {
    Serial.println("Attempting MQTT connection...");
    if (client.connect(MQTT_CLIENT_NAME)) {
      Serial.println("Connected");
    } else {
      Serial.print("Failed, rc=");
      Serial.print(client.state());
      delay(2000);
    }
  }
}

void normalizeData(uint32_t* data, uint32_t length) {
  uint32_t maxVal = 0;
  for (uint32_t i = 0; i < length; i++) {
    if (data[i] > maxVal) {
```

```
            maxVal = data[i];
        }
    }
    if (maxVal == 0) return; // Prevent division by zero

    for (uint32_t i = 0; i < length; i++) {
        data[i] = (data[i] * 1000) / maxVal; // Normalizing to a max value of 1000
    }
}

void freqDetec() {
    if (countData == MAX_BUFFER) {
        if (prevData[roundrobin] < avgData * 1.5 && newData >= avgData * 1.5) { // increasing
and crossing last midpoint
            period = millis() - millistimer; // get period from current timer value
            millistimer = millis(); // reset timer
            maxData = 0;
        }
    }
    roundrobin++;
    if (roundrobin >= MAX_BUFFER) {
        roundrobin = 0;
    }
    if (countData < MAX_BUFFER) {
        countData++;
        sumData += newData;
    } else {
        sumData += newData - prevData[roundrobin];
    }
    avgData = sumData / countData;
    if (newData > maxData) {
```

```
      maxData = newData;
    }

    prevData[roundrobin] = newData; // store previous value

    // Normalize data before printing
    normalizeData(prevData, countData);

    Serial.print(newData);
    Serial.print("\t");
    Serial.print(avgData);
    Serial.print("\t");
    Serial.print(avgData * 1.5);
    Serial.print("\t");
    Serial.print(maxData);
    Serial.print("\t");
    Serial.println(beatspermin);
}

void setup() {
    Serial.begin(115200);
    pinMode(led1, OUTPUT);
    pinMode(led2, OUTPUT);
    pinMode(buzzer, OUTPUT);

    WiFi.begin(WIFISSID, PASSWORD);
    display.begin(SSD1306_SWITCHCAPVCC, SCREEN_ADDRESS);
    display.clearDisplay();
    display.setTextSize(2);
    display.setTextColor(SSD1306_WHITE);
    display.setCursor(0, 0);
```

```
  display.println("Heart Rate Monitor");
  display.println("Waiting for WiFi");
  display.display();

  Serial.print("Connecting to WiFi");
  while (WiFi.status() != WL_CONNECTED) {
    Serial.print(".");
    delay(500);
  }
  Serial.println("\nWiFi connected");
  Serial.println("IP address: ");
  Serial.println(WiFi.localIP());

  client.setServer(mqttBroker, 1883);
  client.setCallback(callback);

  if (!SD.begin()) {
    Serial.println("Card Mount Failed");
    return;
  }
  if (SD.cardType() == CARD_NONE) {
    Serial.println("No SD card attached");
    return;
  }
  createDir(SD, "/mydir");
  writeFile(SD, "/HeartMonitorLog.txt", "University of Ghana: Log of Data\n");
  appendFile(SD, "/HeartMonitorLog.txt", "Holter Monitor\n");
}

void createDir(fs::FS &fs, const char *path) {
  Serial.printf("Creating Dir: %s\n", path);
```

```
  if (fs.mkdir(path)) {
    Serial.println("Dir created");
  } else {
    Serial.println("mkdir failed");
  }
}

void loop() {
  newData = analogRead(SENSOR);
  t1Callback();
  mqttthings();
  freqDetec();

  if (period != lastperiod) {
    frequency = 1000 / (double)period; // timer rate/period
    if (frequency * 60 > 20 && frequency * 60 < 200) { // suppress unrealistic data
      beatspermin = frequency * 60;
      lastperiod = period;
    }
  }
  delay(50);
}

void readFile(fs::FS &fs, const char *path) {
  Serial.printf("Reading file: %s\n", path);
  File file = fs.open(path);
  if (!file) {
    Serial.println("Failed to open file for reading");
    return;
  }
  while (file.available()) {
```

```
      Serial.write(file.read());
    }
    file.close();
}

void writeFile(fs::FS &fs, const char *path, const char *message) {
    Serial.printf("Writing file: %s\n", path);
    File file = fs.open(path, FILE_WRITE);
    if (!file) {
      Serial.println("Failed to open file for writing");
      return;
    }
    if (file.print(message)) {
      Serial.println("File written");
    } else {
      Serial.println("Write failed");
    }
    file.close();
}

void appendFile(fs::FS &fs, const char *path, const char *message) {
    File file = fs.open(path, FILE_APPEND);
    if (!file) {
      Serial.println("Failed to open file for appending");
      return;
    }
    if (!file.print(message)) {
      Serial.println("Append failed");
    }
    file.close();
}
```

```
void renameFile(fs::FS &fs, const char *path1, const char *path2) {
  Serial.printf("Renaming file %s to %s\n", path1, path2);
  if (fs.rename(path1, path2)) {
    Serial.println("File renamed");
  } else {
    Serial.println("Rename failed");
  }
}

void t1Callback() {
  if (millis() - tsLastReport > REPORTING_PERIOD_MS) {
    float myecg = analogRead(SENSOR);
    dtostrf(myecg, 4, 2, str_val_1);
    dtostrf(beatspermin, 4, 2, str_val_2);

    // Update display
    display.clearDisplay();
    display.setTextSize(1);
    display.setTextColor(SSD1306_WHITE);
    display.setCursor(0, 0);
    display.println("HRm:");
    display.setCursor(30, 0);
    display.println(str_val_1);
    display.setCursor(0, 20);
    display.println("BPM:");
    display.setCursor(30, 20);
    display.println(str_val_2);
    display.display();

    // Append to SD card
```

```
      sprintf(payload, "{\"%s\": %s, \"%s\": %s}", VARIABLE_LABEL_1, str_val_1,
VARIABLE_LABEL_2, str_val_2);
      appendFile(SD, "/HeartMonitorLog.txt", payload);
      appendFile(SD, "/HeartMonitorLog.txt", "\n");

      tsLastReport = millis();
    }
}

void mqttthings() {
    if (!client.connected()) {
      reconnect();
    }

    dtostrf(analogRead(SENSOR), 4, 2, str_val_1);
    dtostrf(beatspermin, 4, 2, str_val_2);

    // Publish heart rhythm
    sprintf(topic, "ug/holter/heart_rhythm");
    sprintf(payload, "%s", str_val_1);
    client.publish(topic, payload);

    // Publish pulse (BPM)
    sprintf(topic, "ug/holter/pulse");
    sprintf(payload, "%s", str_val_2);
    client.publish(topic, payload);

    client.loop();
}
```

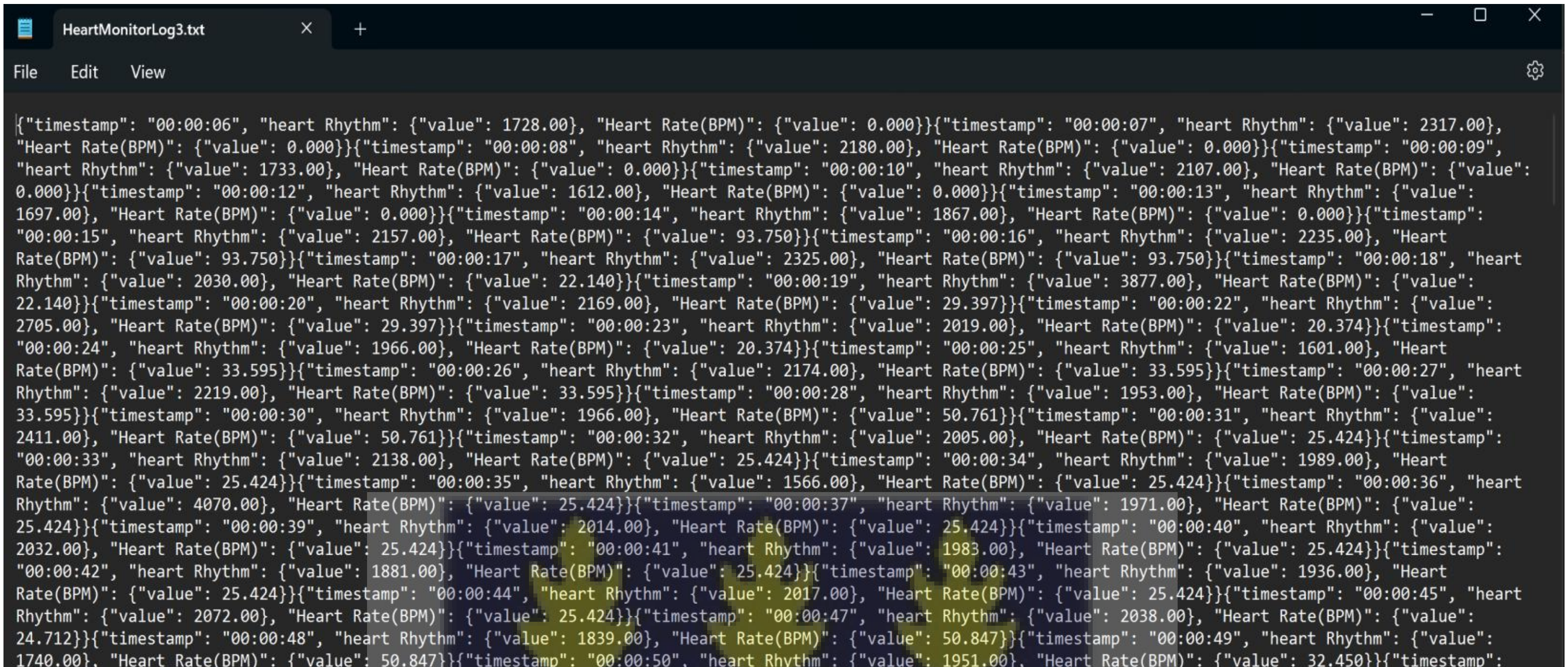
APPENDIX H: SOURCE CODE FOR PLOTTING HEART RATE AND HEART RHYTHM GRAPHS

This was done using Google Colab which is a cloud-based platform that provides free access to computing resources as well as pre-installed libraries for machine learning and data analysis.

```
import json
import matplotlib.pyplot as plt
import numpy as np
from scipy.interpolate import interp1d
# File Directory
file_path = '/content/HeartMonitorLog.txt'
# Reading data from the uploaded file
with open(file_path, 'r') as file:
   # Read the whole file content
   data_string = file.read()
# Adding commas between JSON objects to create a list of objects
data_list = f"[{data_string.replace('}{', '},{')}]"
# Save the modified content to a new file
new_file_path = '/content/ModifiedHeartMonitorLog.txt'
with open(new_file_path, 'w') as new_file:
   new_file.write(data_list)
# Load the data as a list of JSON objects
json_data = json.loads(data_list)
# Extracting heart rhythm and heart rate values
heart_rhythm_values = [entry["heart Rhythm"]["value"] for entry in json_data]
heart_rate_values = [entry["Heart Rate(BPM)"]["value"] for entry in json_data]
# Generating new indices for the interpolated data
new_indices = np.linspace(0, len(heart_rhythm_values) - 1, 10 * len(heart_rhythm_values))
# Interpolating heart rhythm values
interp_func_rhythm = interp1d(range(len(heart_rhythm_values)), heart_rhythm_values,
kind='cubic')
interpolated_rhythm = interp_func_rhythm(new_indices)
# Plotting interpolated heart rhythm and raw heart rate values
plt.figure(figsize=(20, 5))
plt.subplot(1, 2, 1)
plt.plot(new_indices, interpolated_rhythm, linestyle='-', color='red')
plt.title('Interpolated Heart Rhythm')
plt.xlabel('Data Points')
plt.ylabel('Heart Rhythm')
plt.grid(True)
plt.subplot(1, 2, 2)
plt.plot(range(len(heart_rate_values)), heart_rate_values, linestyle='-', color='blue')
plt.title('Raw Heart Rate')
plt.xlabel('Data Points')
plt.ylabel('Heart Rate (BPM)')
```

```
plt.grid(True)
plt.tight_layout()
plt.show()
```

APPENDIX I: SAMPLE DATA OF HEART RYTHYM AND HEART RATE ON SD CARD



{"timestamp": "00:00:06", "heart Rhythm": {"value": 1728.00}, "Heart Rate(BPM)": {"value": 0.000}}{"timestamp": "00:00:07", "heart Rhythm": {"value": 2317.00}, "Heart Rate(BPM)": {"value": 0.000}}{"timestamp": "00:00:08", "heart Rhythm": {"value": 2180.00}, "Heart Rate(BPM)": {"value": 0.000}}{"timestamp": "00:00:09", "heart Rhythm": {"value": 1733.00}, "Heart Rate(BPM)": {"value": 0.000}}{"timestamp": "00:00:10", "heart Rhythm": {"value": 2107.00}, "Heart Rate(BPM)": {"value": 0.000}}{"timestamp": "00:00:12", "heart Rhythm": {"value": 1612.00}, "Heart Rate(BPM)": {"value": 0.000}}{"timestamp": "00:00:13", "heart Rhythm": {"value": 1697.00}, "Heart Rate(BPM)": {"value": 0.000}}{"timestamp": "00:00:14", "heart Rhythm": {"value": 1867.00}, "Heart Rate(BPM)": {"value": 0.000}}{"timestamp": "00:00:15", "heart Rhythm": {"value": 2157.00}, "Heart Rate(BPM)": {"value": 93.750}}{"timestamp": "00:00:16", "heart Rhythm": {"value": 2235.00}, "Heart Rate(BPM)": {"value": 93.750}}{"timestamp": "00:00:17", "heart Rhythm": {"value": 2325.00}, "Heart Rate(BPM)": {"value": 93.750}}{"timestamp": "00:00:18", "heart Rhythm": {"value": 2030.00}, "Heart Rate(BPM)": {"value": 22.140}}{"timestamp": "00:00:19", "heart Rhythm": {"value": 3877.00}, "Heart Rate(BPM)": {"value": 22.140}}{"timestamp": "00:00:20", "heart Rhythm": {"value": 2169.00}, "Heart Rate(BPM)": {"value": 29.397}}{"timestamp": "00:00:22", "heart Rhythm": {"value": 2705.00}, "Heart Rate(BPM)": {"value": 29.397}}{"timestamp": "00:00:23", "heart Rhythm": {"value": 2019.00}, "Heart Rate(BPM)": {"value": 20.374}}{"timestamp": "00:00:24", "heart Rhythm": {"value": 1966.00}, "Heart Rate(BPM)": {"value": 20.374}}{"timestamp": "00:00:25", "heart Rhythm": {"value": 1601.00}, "Heart Rate(BPM)": {"value": 33.595}}{"timestamp": "00:00:26", "heart Rhythm": {"value": 2174.00}, "Heart Rate(BPM)": {"value": 33.595}}{"timestamp": "00:00:27", "heart Rhythm": {"value": 2219.00}, "Heart Rate(BPM)": {"value": 33.595}}{"timestamp": "00:00:28", "heart Rhythm": {"value": 1953.00}, "Heart Rate(BPM)": {"value": 33.595}}{"timestamp": "00:00:30", "heart Rhythm": {"value": 1966.00}, "Heart Rate(BPM)": {"value": 50.761}}{"timestamp": "00:00:31", "heart Rhythm": {"value": 2411.00}, "Heart Rate(BPM)": {"value": 50.761}}{"timestamp": "00:00:32", "heart Rhythm": {"value": 2005.00}, "Heart Rate(BPM)": {"value": 25.424}}{"timestamp": "00:00:33", "heart Rhythm": {"value": 2138.00}, "Heart Rate(BPM)": {"value": 25.424}}{"timestamp": "00:00:34", "heart Rhythm": {"value": 1989.00}, "Heart Rate(BPM)": {"value": 25.424}}{"timestamp": "00:00:35", "heart Rhythm": {"value": 1566.00}, "Heart Rate(BPM)": {"value": 25.424}}{"timestamp": "00:00:36", "heart Rhythm": {"value": 4070.00}, "Heart Rate(BPM)": {"value": 25.424}}{"timestamp": "00:00:37", "heart Rhythm": {"value": 1971.00}, "Heart Rate(BPM)": {"value": 25.424}}{"timestamp": "00:00:39", "heart Rhythm": {"value": 2014.00}, "Heart Rate(BPM)": {"value": 25.424}}{"timestamp": "00:00:40", "heart Rhythm": {"value": 2032.00}, "Heart Rate(BPM)": {"value": 25.424}}{"timestamp": "00:00:41", "heart Rhythm": {"value": 1983.00}, "Heart Rate(BPM)": {"value": 25.424}}{"timestamp": "00:00:42", "heart Rhythm": {"value": 1881.00}, "Heart Rate(BPM)": {"value": 25.424}}{"timestamp": "00:00:43", "heart Rhythm": {"value": 1936.00}, "Heart Rate(BPM)": {"value": 25.424}}{"timestamp": "00:00:44", "heart Rhythm": {"value": 2017.00}, "Heart Rate(BPM)": {"value": 25.424}}{"timestamp": "00:00:45", "heart Rhythm": {"value": 2072.00}, "Heart Rate(BPM)": {"value": 25.424}}{"timestamp": "00:00:47", "heart Rhythm": {"value": 2038.00}, "Heart Rate(BPM)": {"value": 24.712}}{"timestamp": "00:00:48", "heart Rhythm": {"value": 1839.00}, "Heart Rate(BPM)": {"value": 50.847}}{"timestamp": "00:00:49", "heart Rhythm": {"value": 1740.00}, "Heart Rate(BPM)": {"value": 50.847}}{"timestamp": "00:00:50", "heart Rhythm": {"value": 1951.00}, "Heart Rate(BPM)": {"value": 32.450}}{"timestamp":

Figure 48. Sample data of heart rhythm and heart rate stored on the SD card

APPENDIX J: VARIOUS SIMULATIONS CARRIED OUT

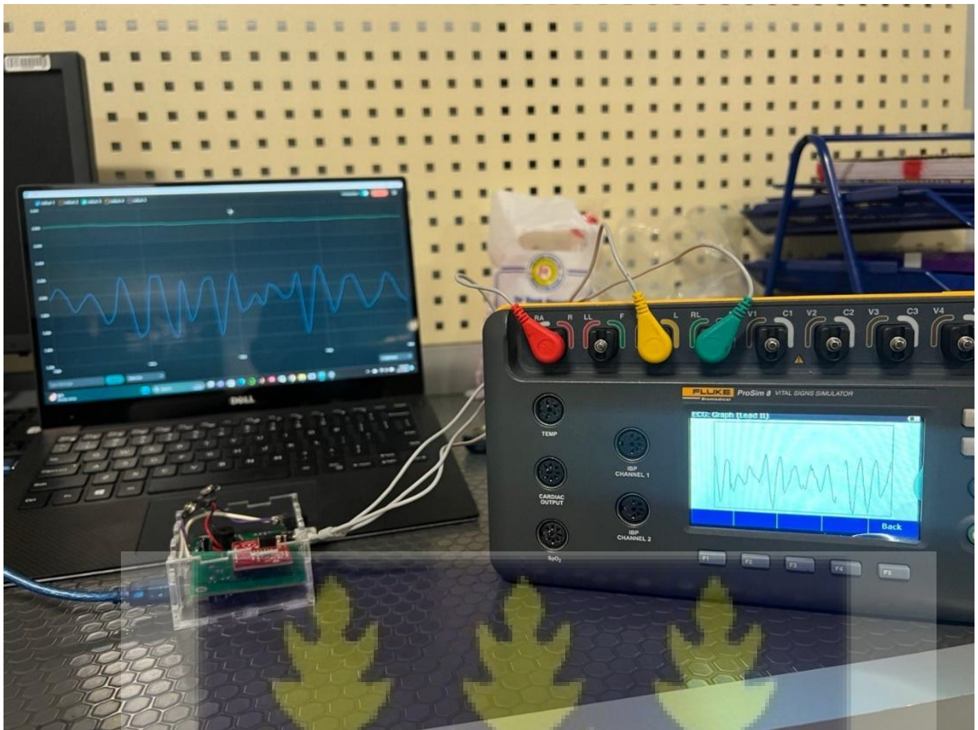


Figure 50. A setup showing a simulated Ventricular Fibrillation graph displayed on the Serial Plotter and the Monitor of the ECG Simulator

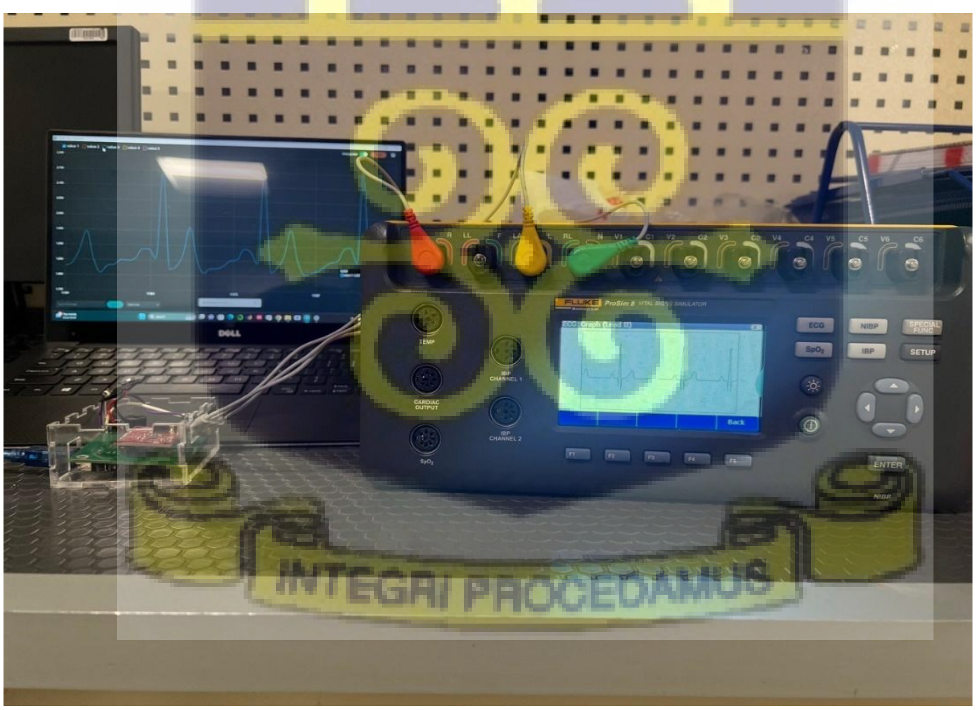


Figure 49. A setup showing a simulated Supraventricular Atrial Tachycardia heart rhythm displayed on the Serial Plotter and the Monitor of the ECG Simulator

Figure 51. A setup showing a simulated Sinus Rhythm with Baseline Wander displayed on the Serial Plotter and the Monitor of the ECG Simulator

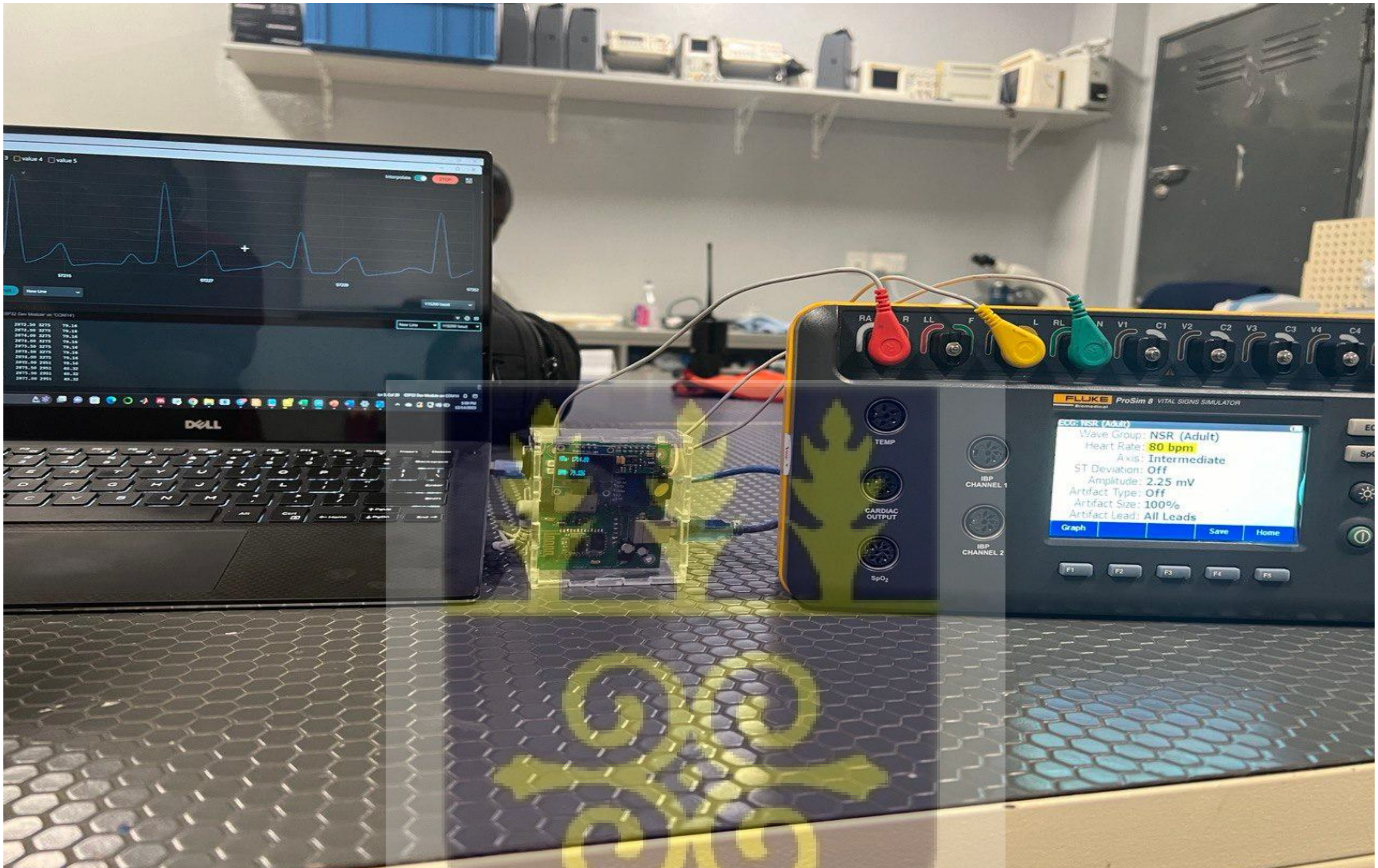


Figure 52. A set up of a simulate Normal Sinus Rhythm at 80bpm with heart beat displayed on the serial monitor and on the Holter Monitor

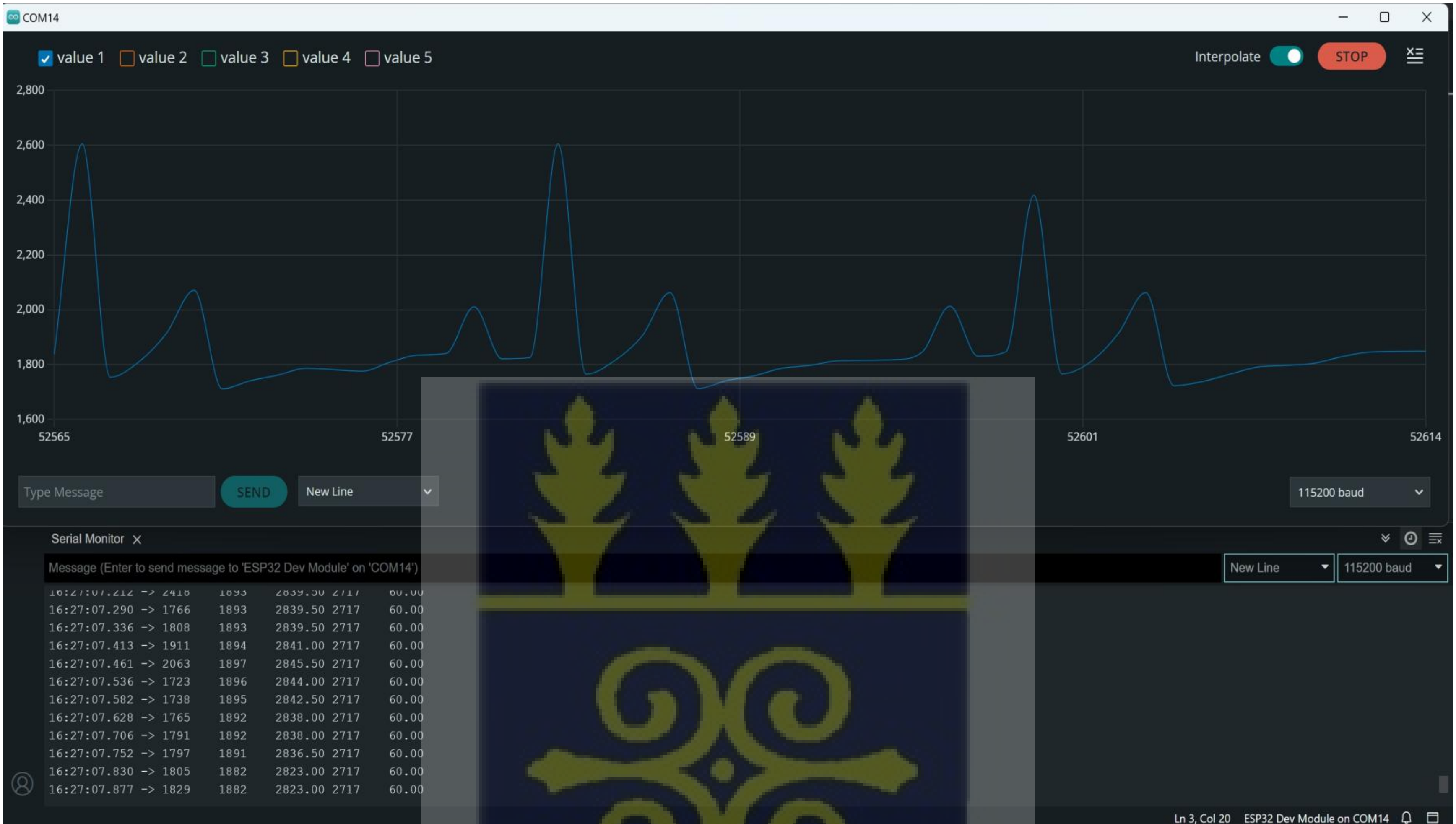


Figure 53. This shows a simulated Sinus Heart Rhythm at 60bpm displayed on the serial plotter

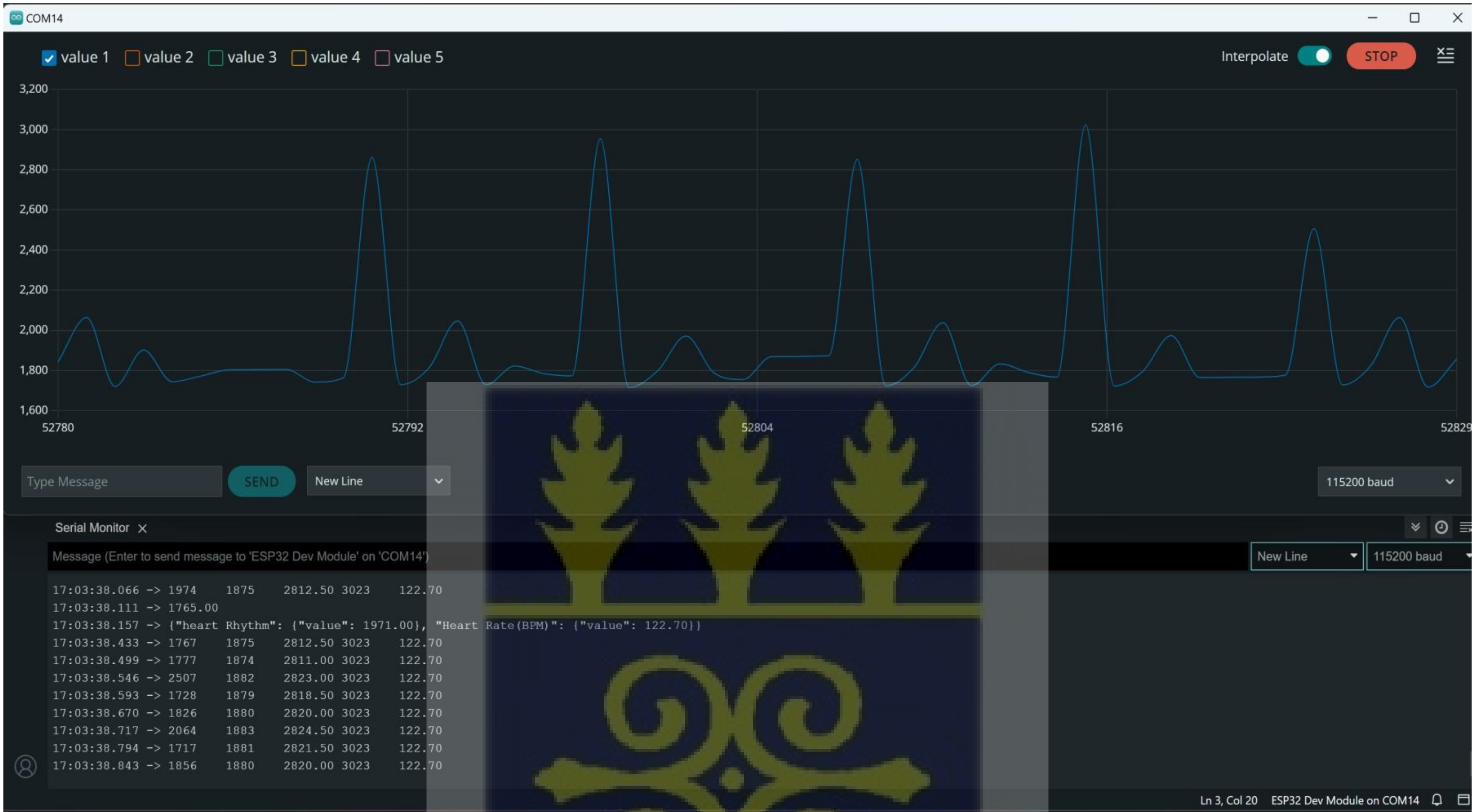
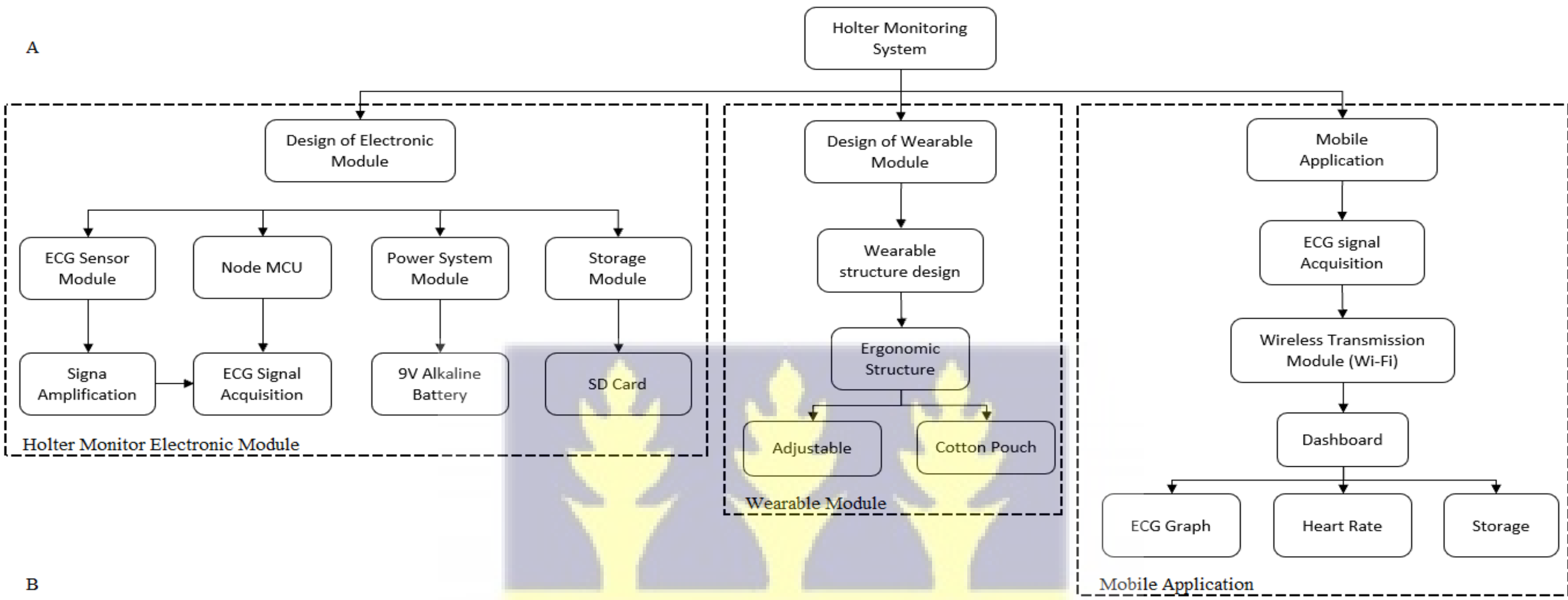


Figure 54. This shows a simulated Sinus Heart Rhythm at 120bpm displayed on the serial plotter

APPENDIX K: OVERALL DESIGN OVERVIEW

A



B