CONTACTLESS MECHANOCARDIOGRAPHY

by

Faranak Mohammad-Zadeh
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APPROVAL

Name: Faranak Mohammad-Zadeh
Degree: Master of Engineering Science
Title of Thesis: Contactless Mechanocardiography

Examining Committee:

Chair: Dr. Parvaneh Saedie
Assistant professor – School of Engineering Science - SFU

Dr. Bozena Kaminska
Senior Supervisor
Professor, Canada Research Chair (Tier 1) in Wireless Sensor Networks

Dr. Carlo Menon
Assistant Professor – School of Engineering Science - SFU

Dr. Pawel Gburzynski
Internal Examiner
Visiting Professor – Simon Fraser University
Professor – University of Alberta

Date Defended/Approved: April 04, 2008
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ABSTRACT

The contactless mechanocardiography (MCG) device is a hand-held contactless heart monitoring device to be used for heart monitoring at home, in hospitals and in other secured environments. Cardiovascular diseases (CD) are currently the first leading cause of death in North America, and the elderly population of British Columbia is increasing rapidly, so we strongly believe that this device can help improve overall health care monitoring. This apparatus can detect the subject's heart signal without any contact with the subject's body. At a distance of up to 30 cm, this device is capable of detecting the heart's signal. It can operate through a mattress, wall or other barriers. The received vital sign signal is simultaneously shown on the device's display.

Keywords: Heart monitoring, vital signs, radar, mechanocardiography (MCG)
EXECUTIVE SUMMARY

Cardiovascular diseases (CD) are the leading cause of death in North America and it is proven that monitoring the heart’s mechanical signal can improve the chances for early detection of heart disease. Consequently, the main objective of this thesis is to outline a device that we have designed to achieve the goal of earlier detection of cardiac abnormalities.

While many tools have been designed to monitor the heart, most of them require contact with the body (electrocardiography, or ECG, is one example). Royal Philips Electronics (a global leader in health care) has investigated and patented a methodology based on a 2.45 GHz radar sensor (KMY24) to detect the heart’s signal. They synchronized the signal taken from the heart with an impedance cardiography (ICG) signal in order to extract the information from the signal that is detected. In their experiments, they have connected the sensor to the subject’s chest [46].

The “contactless mechanocardiography device”, which we have designed and outline in this thesis, requires no contact with the body and is capable of detecting the heart’s signal at a distance of up to 30 cm and sending the information to the medical personnel via cell phone, PDA or a personal computer.

Furthermore, the contactless mechanocardiography device records the heart’s mechanical signal on a real-time basis. The device is equipped with a display that makes it capable of showing the heart's signal on-screen, so there is no need for an oscilloscope or a computer to be used as a display. While the device focuses specifically on the heart signal, its ultimate purpose is to improve the overall efficiency of the health care system.
DEDICATION

“To my parents, who have loved me unconditionally.”
ACKNOWLEDGEMENTS

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## GLOSSARY

<table>
<thead>
<tr>
<th>Term</th>
<th>Definition</th>
</tr>
</thead>
<tbody>
<tr>
<td>Mechanocardiography</td>
<td>Measuring any mechanical behaviour of the heart is called 'mechanocardiography'.</td>
</tr>
<tr>
<td>Doppler Radar</td>
<td>Doppler radar uses the Doppler effect to measure the radial velocity of targets in the antenna's directional beam. The Doppler effect shifts the received frequency up or down based on the radial velocity of target (closing or opening) in the beam, allowing for the direct and highly accurate measurement of target velocity.</td>
</tr>
<tr>
<td>Electrocardiogram (ECG)</td>
<td>ECG (electrocardiogram) is a test that measures the electrical activity of the heart.</td>
</tr>
<tr>
<td>Ballistocardiography (BCG)</td>
<td>The ballistocardiograph (BCG) is a vital sign in the 1-20 Hz frequency range, which is caused by the mechanical movement of the heart and can be recorded by non-invasive methods from the surface of the body. The effect of main heart malfunctions can be identified by observing and analyzing the BCG signal.</td>
</tr>
<tr>
<td>Phonocardiography</td>
<td>The phonocardiogram usually supplements the information obtained by listening to body sounds with a stethoscope (auscultation) and is of special diagnostic value when performed simultaneously with measurement of the electrical properties of the heart (ECG) and pulse rate.</td>
</tr>
<tr>
<td>CW radar</td>
<td>Continuous-wave radar</td>
</tr>
<tr>
<td>PRT</td>
<td>Pulse repetition time</td>
</tr>
<tr>
<td>US</td>
<td>Ultrasound</td>
</tr>
<tr>
<td>MHR</td>
<td>Maximum heart rate</td>
</tr>
<tr>
<td>HRV</td>
<td>Heart rate variability</td>
</tr>
<tr>
<td>IM</td>
<td>Intima-media</td>
</tr>
<tr>
<td>CV/CD</td>
<td>Cerebrovascular disease / cardiovascular disease</td>
</tr>
<tr>
<td>AV</td>
<td>Atrioventricular</td>
</tr>
<tr>
<td>Abbreviation</td>
<td>Definition</td>
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<tr>
<td>--------------</td>
<td>----------------------------------</td>
</tr>
<tr>
<td>SA</td>
<td>Sinoatrial</td>
</tr>
<tr>
<td>HR</td>
<td>Heart rate</td>
</tr>
<tr>
<td>RVSM</td>
<td>Radar Vital Signs Monitor</td>
</tr>
<tr>
<td>USB</td>
<td>Universal Serial Bus</td>
</tr>
<tr>
<td>UART</td>
<td>Universal Asynchronous Receiver/Transmitter</td>
</tr>
<tr>
<td>NiCd</td>
<td>Nickel Cadmium</td>
</tr>
<tr>
<td>LP</td>
<td>Low pass filter</td>
</tr>
<tr>
<td>HP</td>
<td>High pass filter</td>
</tr>
<tr>
<td>MRT</td>
<td>Magnet Resonance Tomography</td>
</tr>
</tbody>
</table>
CHAPTER 1: INTRODUCTION

Most Western countries face high and increasing rates of cardiovascular disease. Each year, heart disease kills more North Americans than cancer [1]. In North America, diseases of the heart alone cause 30% of all deaths, with other diseases of the cardiovascular system causing substantial further death and disability. Diseases of the heart are the number one cause of both death and disability in the United States and most European countries. A large histological study called PDA Y (The Pathobiological Determinants of Atherosclerosis in Youth) showed vascular injury accumulates from adolescence, making primary prevention efforts necessary from childhood [2][3].

According to PDA Y, the earlier chronic ischemic heart disease is detected, the better are the chances for a treatment before life threatening symptoms occur. Angina pectoris, commonly known as ‘angina’ and ‘heart attack’, are only the last stages of the worsening disease. About 80% of patients show changes in the motion pattern of the heart wall in stages earlier than when ECG artefacts are visible. In a clinical environment, these wall motion artefacts can be observed by imaging technologies like ultrasound (US) or Magnet Resonance Tomography (MRT).
Today, there are more options available for monitoring heart failure than ever before. Careful monitoring is the first step in the prevention of disease. The goals of treating heart failure are primarily to decrease the likelihood of disease progression (thereby decreasing the risk of death and the need for hospitalization), to lessen symptoms and to improve quality of life.

It is important to note that wall motion analysis can detect ventricular dysfunction earlier than is possible with conventional echocardiography [20]. Its usefulness for heart problems has already been reported [20] [22] [23].

Therefore, the main objective of this thesis is to describe our design of a device for the monitoring of the mechanical movement of the heart to detect ventricular dysfunction earlier than has been previously possible. In this device, radar technology plays the main role. This thesis demonstrates that radar technology can be considered as an option for non-invasive and contactless heart monitoring and other forms of mechanocardiography analysis.

The most commonly known application of radar is its use in nautical and aeronautical navigation. Airplanes use radar to detect obstacles, while ships use it for early detection of other vessels and shorelines. It is important to note that radar functions independent of weather. So, an important application of radar is collision detection in poor sight conditions. Radar systems are also used by harbours and airports for sea and
airspace surveillance, respectively. The speed of vehicles can be measured using radar technology as well. Radar is used in the context of road traffic, especially for speed control. Radar can be installed in cars as part of an anti-collision system, as it can measure the distance to the vehicle ahead. An important application of radar for military use is airspace surveillance. In military situations, radar can help distinguish between friends and enemies. Other applications of radar include the guidance of weapons and intelligence services. Radar is applied in numerous different fields, including medicine, and its usage is growing rapidly [39].

A number of research groups have conducted research related to the use of radar technology for medical purposes. The Georgia Institute of Technology, for example, has developed the Radar Vital Signs Monitor (RVMS) device, which uses a very high frequency of 24 and 35 GHz. The system's original objective was to assess a soldier's life signs at a distance of 100 m. Other applications include the surveillance of burn victims, telemedicine for elderly persons and the evaluation of athletes' performance. A parabolic antenna measuring 60 cm in diameter was used in their proposed device (RVMS) [42].

Related research has also been conducted at the University of Rome 'Tor Vergata' and at the Moscow Aviation Institute (MAI). In this case, the researchers have used an Ultra Wide Band Radar (UWB) approach for contactless monitoring of vital signs, but they have not
presented a working system. Their system uses pulsed wave UWB radar with a center frequency of 5 GHz and pulses of 0.05 to 30 MHz [43, 44].

The basis of our research is the work done by Philips Electronics Ltd. They have investigated and patented a methodology based on a 2.45 GHz radar sensor (KMY24) to detect the heart's signal. In order to extract the information from the signal that is detected by the 2.45 GHz sensor (KMY24), they synchronized the signal received from the heart with an impedance cardiography (ICG) signal. In their experiments, Phillips connected the sensor to the chest of the subject so therefore it is not a contactless device [46]. To our knowledge, these examples from the literature have not progressed past the research stage and merely show the basic principle at work.

The objective of our study has been to advance research in the area of contactless heart activity analysis and to explore the use of the same mechanical principles for blood flow determination. We have achieved this aim by using a test methodology and by designing a device that enables not only diagnosis, but also allows for the monitoring of the heart and major blood vessels, with no connection to the subject's body. The device is easy to use anywhere, both in the hospital setting, as well as at home, in secured locations such as airports, in the world of athletics and so on. The device can prove the correspondence of our acquired signal to sternal BCG and phonocardiograph signals.

In this thesis, the following two basic terms are used as follows:
The first term, ‘mechanocardiography’, means measuring any mechanical behaviour of the heart. Correspondingly, measuring any electrical behaviour of the heart (beating or still) is ‘electrocardiography’.

Ballistocardiography, seimocardiography or displacement cardiography are all subsets of mechanocardiography. Therefore, a contactless mechanocardiography (MCG) device detects the mechanical signal of the heart and the MCG signal shows the motion pattern of the heart.

Our approach and methodology can be summarized as follows: One, analyze the background information necessary to understand the electronic system; two, test and compare the results taken from the contactless MCG device; three, gain an understanding of the software developed for heart rate calculation; four, study results and validate those results with other methods; and five, study possible major applications of the contactless MCG device.

The contactless MCG device we have developed consists of a transmitter, receiver, band-pass filter, and an amplifier. The output from the filter and the amplifier is sent to the A/D and then to the microprocessor for further processing. After that, the data is sent to the device’s display.

The device is capable of detecting the heart’s signal at a distance of up to 30 cm, and is powered by 2 AA batteries (2.4 V). To make it a stand-alone device, it is equipped with a display that renders it capable of
displaying the heart's signal on-screen, eliminating the need for an oscilloscope or computer connection to view the signal. The system can be used for quick monitoring or for clinical analysis. Communication between the contactless MCG device and other devices, like a PDA, cell phone or personal computer can be done on a real-time basis.

We have chosen to use the monitoring of heart rate (HR) as an example of how the contactless mechanocardiography device can be directly applied in the biomedical field. HR is a basic vital sign but one that is critical for physiological monitoring. (Figure 1.1 shows a female subject checking her foetus's mechanocardiography signal using a contactless mechanocardiography device by pointing the device toward her belly. This is but one example of the device's application.)

The thesis itself is organized as follows. In the second chapter we review the relevant medical concepts and measuring techniques, such as electrocardiography (ECG) and ballistocardiography (BCG). In Chapter 3, we review the theoretical components of our research. In Chapter 4, the electronic circuit design of the contactless mechanocardiography device is fully explained, as is the controlling software for the microprocessor and the LCD. The basic algorithm for heart-rate calculation is also presented. Finally, in Chapter 5, we show and discuss the results taken from our experimentation with the contactless MCG device at Burnaby Hospital, SFU's School of Kinesiology and at SFU's CIBER lab. The experimental
The study was granted two ethics approvals, one from SFU and one from Burnaby Hospital (Fraser Health Authority).

Figure 1.1. Female subject is checking her foetus's mechanical heart signal by standing in front of the "contactless mechanocardiography device"
CHAPTER 2: MEDICAL BACKGROUND

This chapter reviews the medical background related to our research. Here, we explain the physiology and anatomy of the heart, as well as the thorax and major arteries. We also introduce the concepts of electrocardiography (ECG) and ballistocardiography (BCG), and present the differences among the various mechanocardiography (MCG) methods. We use the heart rate as our main application example. Information about heart rate, and different methods of calculating it, are discussed in section 2.5. Afterwards, we review two heart rate abnormalities, bradycardia and tachycardia.

2.1. Physiology and anatomy of the heart, thorax, and arteries

The heart is an organ the size of a fist, with a somewhat pointed end that is directed towards the left foot (Figure 2.1). At the top are the two thin walled pumps called the right and left atria, which prime the main pumps of the heart, the right and left ventricles. A thick muscular wall called the “septum” separates the two ventricles internally. Blood from the right ventricle goes to the lungs, and from the left ventricle to the rest of the body. The front part of the heart is close to the complex of muscle, rib,
and skin that forms the rib cage. The bottom of the ventricles rests on the
diaphragm, a muscular sheet. To the right and left of the heart is lung
tissue, behind it the esophagus, while beneath the diaphragm is the
stomach, usually with air bubbles, and the liver. Blood vessels, veins and
arteries extend from the heart both upwards and downwards (Figure 2.2).
All fibres of heart muscle undergo spontaneous excitation. The fastest
beating fibres, however, are in the auricular muscle and they are known as
the sinoatrial (SA) node. The SA node is one of the major elements in the
cardiac conduction system. The cardiac conduction system generates
electrical impulses and conducts them throughout the muscle of the heart,
stimulating the heart to contract and pump blood. The SA node is
essentially the pacemaker for the heart.

Nerves release chemicals affecting the heart rate and the strength
of the resulting muscle contraction. The SA node is the source of an
excitation wave that progresses outward through the muscle at a speed of
1 m/s covering the atria in about one second, and producing the P wave of
the ECG (Figure 2.5). After the wave of excitation has passed over a heart
cell, the membrane potential is greatly reduced in magnitude for about 1/3
of a second. The wave of excitation proceeds from the atria to the
ventricles via the atrioventricular node (AV), the top node of the septum,
where it is delayed by about 1/20 of a second. Then, it is transmitted to
various regions in the septum and to the inner ("endocardial") walls of the
ventricles via the “Purkinje system” (a special network of specialized heart cells). This process takes a few milliseconds.

Next, the excitation wave goes radially from the inner ventricles’ wall to the epicardium (the outer ventricles’ wall). As the right ventricle’s wall is about 1/3 as thick as the left ventricle’s wall, this traversal takes less time for the right ventricle’s wall. The passage of the excitation wave into the septum and through the ventricles and the electrical recovery of atrial muscle are associated with the QRS complex of the electrocardiogram (Figure 2.5). In normal subjects, the QRST lasts about 0.1 second. Then, after about 0.2 second, electrical recovery of the ventricle commences. The recovery process takes about 1/3 of a second and is associated with the electrocardiographic T wave and ventricular relaxation. A detailed description of an ECG waveform is provided in section 2.2.

Figure 2.1. A view of the heart’s position in the body
A second key form of vascular disease is ‘cerebrovascular disease’ (CD). Atherosclerotic vascular disease (hardening of the arteries) is a progressive disease caused by a slow build-up of cholesterol deposits (plaque) within the arteries and is a major concern for aging patients. The deposit of plaque that forms on the artery walls (Figure 2.3) can cause a significant decrease of blood flow to muscles and vital organs. When this condition is present in the carotid arteries, there is a much higher risk of stroke. This is a treatable and surgically correctable condition [29].
Figure 2.3. A diseased artery [45]

The carotid arteries are the major blood vessels that supply the head and brain with the necessary blood flow to deliver oxygen. They branch off the aorta (the main artery leaving the heart) with some variation between the right and left side of the body, and travel up the throat, one on each side of the windpipe (right and left common carotid arteries). At about the jawbone, there is another branching into the external and internal carotid arteries. The external carotid supplies blood to the face, scalp and other external head tissues. The internal carotid supplies blood to the brain. It is at this branching that most plaque build-up occurs and, therefore, where most blockages are found. The major concern is the appropriate blood flow through the internal carotid arteries as they supply the brain (Figure 2.4). In most studies, intima-media (IM) of the arteries'
wall thickness has been associated with risk factors for cardiovascular disease [30 - 37].

2.2. Electrocardiography (ECG)

The action potential created by heart wall contraction spreads electrical currents from the heart throughout the body. The spreading electrical currents create different potentials at different points in the body, which can be sensed by electrodes on the skin surface using biological transducers made of metals and salts. This electrical potential is an AC signal with a bandwidth of 0.05 Hz to 100 Hz, sometimes up to 1 kHz. It is generally around 1-mV peak-to-peak in the presence of much larger external high frequency noise plus 50-/60-Hz interference normal-mode
(mixed with the electrode signal) and common-mode voltages (common to all electrode signals).

Figure 2.5 shows a typical ECG signal. The small initial waveform, the P wave, stems from the electrical activation of the atrium and is about 90 ms in duration. The large peaked deflection, the QRS complex, is due to activation of the ventricles and lasts about 80 ms. The interval between the P wave and the QRS complex is called the atrioventricular delay and lasts about 80 ms. The final, slowly varying wave, the T wave, is related to the electrical recovery of the ventricles. Electrical effects of recovery of the atrium are buried in the QRS complex.

![ECG waveform](image)

**Figure 2.5. Nomenclature of the electrocardiogram [41].**

In Wilson [17] and Wilson, Johnston, Rosenbaum, and Barker [18], it was assumed by the model that the body was part of an infinite bounded
To measure the potential to infinity in a bounded conductor such as the body, Wilson suggested using a “central terminal,” i.e., a terminal connected to the right arm (R), left arm (L), and left leg (F) by equal resistors (Figure 2.6), as an equivalent electrode at infinity. So, potential differences measured between the central terminal and the limb electrodes, or an electrode on the chest surface (exploring electrode) were thought to approximate the potentials which would exist relative to infinity, assuming the body were actually part of an infinite homogeneous conductor.

![Figure 2.6. The standard positions of chest leads](image)

Each ECG lead records the electrical activity of the heart from a different perspective, which also correlates to different anatomical areas of the heart for identifying acute coronary ischemia or injury. Two leads that look at the same anatomical area of the heart are said to be contiguous.
The inferior leads (leads II, III and aVF) look at electrical activity from the vantage point of the inferior (or diaphragmatic) wall of the left ventricle. The lateral leads (I, aVL, V5 and V6) look at the electrical activity from the vantage point of the lateral wall of the left ventricle. Because the positive electrode for leads I and aVL are located on the left shoulder, leads I and aVL are sometimes referred to as the high lateral leads. Because the positive electrodes for leads V5 and V6 are on the patient's chest, they are sometimes referred to as the low lateral leads.

The septal leads, V1 and V2, look at electrical activity from the vantage point of the septal wall of the left ventricle. They are often grouped together with the anterior leads. The anterior leads, V3 and V4, look at electrical activity from the vantage point of the anterior wall of the left ventricle.

In addition, any two precordial leads that are next to one another are considered to be contiguous. For example, even though V4 is an anterior lead and V5 is a lateral lead, they are contiguous because they are next to one another.

Lead aVR offers no specific view of the left ventricle. Rather, it views the inside of the endocardial wall from its perspective on the right shoulder [38].
2.3. Ballistocardiography (BCG) and mechanocardiography (MCG)

BCG stands for ballistocardiography and the ballistocardiograph is a vital signal in the 1-20 Hz frequency range, which is caused by the movement of the heart and blood and can be recorded from the human body by non-invasive means.

As can be seen in Figure 2.7, the ideal BCG waveform consists of seven waveform peaks labelled H through N as defined by the American Heart Association. ‘H’ is the first upward deflection after the electrocardiograph (ECG) R-wave on the acceleration BCG when recorded simultaneously. The letter ‘I’ is the downward wave immediately after ‘H’, and lastly the letter ‘J’ is the upward wave after ‘I’. The ‘L’, ‘M’ and ‘N’ waves correspond to the diastolic phase of the cardiac cycle [4]. In addition to a number of clinical studies that have been performed with BCG, specialized BCG instruments, including beds [9] and chairs [8], have been developed by different research groups [10].

In the early 1930s, Isaac Starr recognized that BCG signals closely reflect the strength of myocardial contraction [4, 5] and classified them into four groups, as can be seen in Figure 2.8. According to Starr’s classification, BCG signals can be divided into four different classes depending on the abnormalities reflected by the measured signals. The Starr classification (Figure 2.8) is done visually, based on the following guidelines. In class 1, all BCG complexes are normal in contour. In class
2, the majority of the complexes are normal, but one or two of the smaller complexes of each respiratory cycle are abnormal in contour. In class 3, the majority of the complexes are abnormal in contour, while usually only a few of the largest complexes of each respiratory cycle remain normal and in class 4, there is such complete distortion that the waves cannot be identified with confidence and the onset of ejection cannot not be located without the assistance of a simultaneous ECG [8].

![Figure 2.7. BCG signal of a single heart-cycle as recorded by the multisensor referenced to synchronized ECG](image)

As a result of his valuable research, clinicians and medical experts for almost three decades studied the effects of different heart malfunctions by means of BCG and proved that these malfunctions can be related to typical patterns in BCG signals [6,7].
A mechanocardiography (MCG) signal shows the mechanical effects of the heartbeat, which means that BCG is in fact a form of MCG signal. The advantage of BCG analysis over ECG analysis is that no electrodes need to be attached to the subject. An ECG does not show information about the mechanical movement pattern of the heart.

Figure 2.8. These are longitudinal BCG records that shows Starr’s method for BCG signal classification [4]

2.4. Thorax

A cross section of the thorax is shown in Figure 2.9 at the height of the heart, i.e. the fifth rib. In the center of the Figure 2.9 the heart is visible, consisting of two ventricles: the thick-walled left ventricle (in the picture on the right) and the thin-walled right ventricle. Below the heart there are the esophagus and major blood vessels. Most of the cross-section is filled by the lungs, surrounded by bones: the ribs, the sternum in
the front (top of picture) and the vertebral column in the back. Further outwards is some muscle tissue, followed by fat and — finally — the skin.

Figure 2.9. Cross section of the human thorax, taken from the Visible Human Project

There are no layers of different tissue between the sternum and the heart so the best position to get the heart’s mechanical signal is around the sternum.

While monitoring the left ventricle of the heart a beam would have to cross the following tissue layers:

- Skin (approximately 2 mm)
- Fat (approximately 10 mm)
- Muscle (approximately 10 mm)
- Bone or cartilage (approximately 8 mm) – depending on whether viewing through or between the ribs
• Lung (approximately 10 mm) – highly dependent on whether inflated or deflated

A shorter way leads to the right ventricle, which is located directly beneath the sternum (Path B in Figure 2.9):

• Skin (approx. 2 mm)
• Fat (approx. 8 mm)
• Muscle (approx. 5 mm)
• Bone: the sternum (approx. 10 mm)

In addition, Table 1 shows that there is a substantial conductivity difference between bone and muscle. Typical wavelengths in the different body tissues are between 1 cm and 4 cm, although in fat, electromagnetic waves have a larger wavelength.

<table>
<thead>
<tr>
<th>Tissue name</th>
<th>Conductivity [S/m]</th>
<th>Relative permittivity</th>
<th>Loss tangent</th>
<th>Wavelength [m]</th>
<th>Penetration depth [m]</th>
</tr>
</thead>
<tbody>
<tr>
<td>Air</td>
<td>0.000</td>
<td>1.000</td>
<td>0.000</td>
<td>9.9931</td>
<td>N/A</td>
</tr>
<tr>
<td>Aorta</td>
<td>0.33247</td>
<td>144.4</td>
<td>13.795</td>
<td>0.0538</td>
<td>0.52253</td>
</tr>
<tr>
<td>Bladder</td>
<td>0.25133</td>
<td>132.48</td>
<td>11.367</td>
<td>0.4853</td>
<td>0.60564</td>
</tr>
<tr>
<td>Blood</td>
<td>0.98268</td>
<td>1080</td>
<td>5.4518</td>
<td>1.6812</td>
<td>0.32111</td>
</tr>
<tr>
<td>Blood Vessel</td>
<td>0.33247</td>
<td>144.4</td>
<td>13.795</td>
<td>0.0538</td>
<td>0.52253</td>
</tr>
<tr>
<td>Body Fluid</td>
<td>1.5015</td>
<td>73.701</td>
<td>122.07</td>
<td>1.4839</td>
<td>0.23811</td>
</tr>
<tr>
<td>Bone Cancellous</td>
<td>0.10256</td>
<td>148.34</td>
<td>4.1424</td>
<td>5.0586</td>
<td>1.0226</td>
</tr>
<tr>
<td>Bone Cortical</td>
<td>0.031856</td>
<td>83.251</td>
<td>2.2928</td>
<td>8.2775</td>
<td>2.0119</td>
</tr>
<tr>
<td>Bone Marrow</td>
<td>0.0055988</td>
<td>30.824</td>
<td>1.0884</td>
<td>16.17</td>
<td>5.8597</td>
</tr>
<tr>
<td>Brain Grey Matter</td>
<td>0.19654</td>
<td>564.72</td>
<td>2.0854</td>
<td>3.2674</td>
<td>0.82609</td>
</tr>
<tr>
<td>Brain White Matter</td>
<td>0.11905</td>
<td>285.39</td>
<td>2.4995</td>
<td>4.3537</td>
<td>1.0235</td>
</tr>
<tr>
<td>Breast Fat</td>
<td>0.02679</td>
<td>14.205</td>
<td>11.3</td>
<td>10.672</td>
<td>1.8555</td>
</tr>
<tr>
<td>Cartilage</td>
<td>0.30406</td>
<td>557.34</td>
<td>3.2689</td>
<td>2.8479</td>
<td>0.61265</td>
</tr>
<tr>
<td>Cerebellum</td>
<td>0.23003</td>
<td>810.45</td>
<td>1.7007</td>
<td>2.8791</td>
<td>0.80101</td>
</tr>
<tr>
<td>Cerebro Spinal Fluid</td>
<td>2.0002</td>
<td>108.96</td>
<td>109.99</td>
<td>1.2851</td>
<td>0.20639</td>
</tr>
</tbody>
</table>
2.5. Heart rate

"Heart rate" is the number of contractions of the heart per minute. It is considered one of the key vital signs. Usually it is calculated as the number of contractions (heartbeats) of the heart in one minute and expressed as "beats per minute" (bpm). Table 2 provides typical heart rates for different age groups.

Table 2. Relationship of the heart rate to age

<table>
<thead>
<tr>
<th>Age</th>
<th>Normal heart rate [24]</th>
</tr>
</thead>
<tbody>
<tr>
<td>Newborn</td>
<td>130</td>
</tr>
<tr>
<td>3 months</td>
<td>150</td>
</tr>
<tr>
<td>6 months</td>
<td>135</td>
</tr>
<tr>
<td>1 year</td>
<td>125</td>
</tr>
<tr>
<td>2 years</td>
<td>115</td>
</tr>
<tr>
<td>3 years</td>
<td>100</td>
</tr>
<tr>
<td>4 years</td>
<td>100</td>
</tr>
<tr>
<td>6 years</td>
<td>100</td>
</tr>
<tr>
<td>8 years</td>
<td>90</td>
</tr>
<tr>
<td>9 years</td>
<td>95</td>
</tr>
<tr>
<td>12 years</td>
<td>85</td>
</tr>
<tr>
<td>Adult</td>
<td>60-100</td>
</tr>
</tbody>
</table>
When resting, the average adult human heart rates are about 70 bpm (males) and 75 bpm (females). However, this rate varies among people and can be significantly lower in athletes. The infant/neonatal rate of heartbeat is around 130-150 bpm, a toddler's about 100–130 bpm, an older child's about 90–110 bpm, and an adolescent's about 80–100 bpm.

Heart rate variability (HRV) is the variation of beat-to-beat intervals. A healthy heart has a large HRV, while decreased or absent variability may indicate cardiac disease. HRV also decreases with exercise-induced tachycardia. HRV has been the focus of increased research, mainly with an eye to its use as a physiological marker to classify different pathological disorders.

One aspect of heart rate variability can be used as a measurement of fitness, specifically the speed at which one's heart rate drops upon termination of vigorous exercise. The speed at which a person's heart rate returns to resting is considerably faster for a fit person than for an unfit person. A drop of 19 beats in a minute is typical for a healthy person. A drop of less than 12 beats per minute after maximal exercise has been correlated with a significant increase in mortality [25].

The pulse rate (which in most people is identical to the heart rate) can be measured at any point on the body where an artery's pulsation is transmitted to the surface - often as it is compressed against an underlying structure like bone. Some commonly palpated sites are as listed:
1. The inside of the wrist on the side of the thumb (radial artery), and less commonly the ulnar artery on the pinky side which is deeper and harder to palpate

2. The neck (carotid artery)

3. The inside of the elbow, or under the bicep muscle (brachial artery)

4. The groin (femoral artery)

5. Behind the medial malleolus on the feet (posterior tibial artery)

6. Middle of dorsum of the foot (dorsalis pedis)

7. Behind the knee (popliteal artery)

8. Over the abdomen (abdominal aorta)

9. The chest (aorta); this can be felt with one's hands or fingers but it is also possible to examine the heart by utilizing a stethoscope [26]

   Maximum heart rate (also called MHR, or HRmax) is the maximum heart rate that a person achieves during maximal physical exertion. Research indicates it is most closely linked to a person's age. A person's HRmax will decline as they age [27]. People who have participated in sports and athletic activities in their early years will have a higher MHR than those less active as children.

   The pulse is the most straightforward way of measuring the heart rate, but it can be deceptive when some heart beats do not have much cardiac output. In these cases (as happens in some arrhythmias), the heart rate may be considerably higher than the pulse rate. Auscultation is also a method of heart rate measurement.
The heart contains two cardiac pacemakers that spontaneously cause the heart to beat. These are controlled by the autonomic nervous system and circulating adrenaline. The vagus nerve (a pneumogastric nerve or cranial nerve X), which governs heart rate, can be controlled through breathing.

2.5.1. Measuring Heart Rate (HR) and HR_{\text{max}}

![Graph showing the relationship between age and heart rate.](image)

**Figure 2.10. Relationship of exercise to heart beat** [48]

There are different methods of measuring the heart rate. Here we explain the "Fox and Haskell formula", the "Karvonen method" and the "Zoladz method".
**Fox and Haskell formula [41]**

The most accurate way of measuring $HR_{\text{max}}$ for an individual is via a cardiac stress test. In such a test, the subject exercises while being monitored by an ECG. During the test, the intensity of exercise is periodically increased (if a treadmill is being used, through increase in speed or slope of the treadmill) until the subject can no longer continue, or until certain changes in heart function are detected in the ECG (at which point the subject is directed to stop). Typical durations of such a test range from 10 to 20 minutes. Since the $HR_{\text{max}}$ declines with age, this test does not hold permanent value.

Conducting an accurate maximal exercise test requires expensive equipment, and should only be performed in the presence of medical staff due to risks associated with high heart rates. Instead, people typically use predictive formulae to estimate their individual maximum heart rate. The most common formula encountered is:

$$HR_{\text{max}} = 220 - \text{age (can vary)} \quad (2.1)$$

This is attributed to various sources, often "Fox and Haskell". While the most common (and easy to remember and calculate), some do not consider this particular formula to be a good predictor of $HR_{\text{max}}$.

A 2003 study [24] of 43 different formulae for $HR_{\text{max}}$ (including the one above) concluded the following:
1) No "acceptable" formula currently existed, (they used the term "acceptable" to mean acceptable for both prediction of $V_{O2max}$, and prescription of exercise training HR ranges)

2) The most accurate formula of those examined was:

$$HR_{max} = 205.8 - (0.685 \times \text{age}) \quad (2.2)$$

This was found to have a standard of error that, although large (6.4 bpm), was still deemed to be acceptable for the use of prescribing exercise training HR ranges.

Other often-cited formulae are:

$$HR_{max} = 206.3 - (0.711 \times \text{age}) \quad (2.3)$$

(Often attributed to "Londeree and Moeschberger from the University of Missouri–Columbia") and

$$HR_{max} = 217 - (0.85 \times \text{age}) \quad (2.4)$$

These figures are still dependent on physiology and fitness - for example, an endurance runner's rates will typically be lower due to the increased size of the heart required to support the exercise, while a sprinter's rates will be higher due to the improved response time and short duration.
The recovery heart rate is that taken 2 to 10 minutes after exercise. It is measured over a 15-second sampling interval. The goal is not to exceed 150 bpm.

A drop of 20 beats in a minute is typical for a healthy person. A drop of less than 12 beats per minute after maximal exercise has been correlated with a significant increase in mortality [26].

Target heart rate (THR), or training heart rate, is a desired range of heart rate reached during aerobic exercise, which enables one's heart and lungs to receive the most benefit from a workout. This theoretical range varies based on one's physical condition, age and previous training. Below are two ways to calculate one's target heart rate. In each of these methods, there is an element called "intensity" which is expressed as a percentage. THR can be calculated by using a range of 50% to 85% intensity.

Karvonen method [41]

The Karvonen method factors in resting heart rate (HR_{rest}) to calculate target heart rate (THR):

$$THR = ((HR_{max} - HR_{rest}) \times \%\text{Intensity}) + HR_{rest} \tag{2.5}$$

An example for someone with a HR_{max} of 180 and a HR_{rest} of 70 would be:
50% intensity: \((180 - 70) \times 0.50) + 70 = 125 \text{ bpm}\)

85% intensity: \((180 - 70) \times 0.85) + 70 = 163 \text{ bpm}\)

**Zoladz method [39]**

An alternative to the Karvonen method is the Zoladz method, which derives exercise zones by subtracting values from \(\text{HRmax}\).

\[
\text{THR} = \text{HRmax} - \text{Adjuster} \pm 5 \text{ bpm} \tag{2.6}
\]

Zone 1 Adjuster = 50 bpm

Zone 2 Adjuster = 40 bpm

Zone 3 Adjuster = 30 bpm

Zone 4 Adjuster = 20 bpm

Zone 5 Adjuster = 10 bpm

An example of someone with a HRmax of 180 would be:

Zone 1 (easy exercise): \(180 - 50 = 130; \pm 5 \rightarrow 125 \text{ to } 135 \text{ bpm}\)

Zone 4 (tough exercise): \(180 - 20 = 160; \pm 5 \rightarrow 155 \text{ to } 165 \text{ bpm}\)

Heart rate reserve (HRR) is a term used to describe the difference between a person’s measured or predicted maximum heart rate and resting heart rate. Some methods for measuring exercise intensity focus on the percentage of heart rate reserve. Additionally, as a person increases their cardiovascular fitness, their \(HR_{\text{rest}}\) will drop; thus, the heart
rate reserve will increase. The percentage of HRR is equivalent to the percentage of VO₂ reserve.

\[ \text{HRR} = \text{HR}_{\text{max}} - \text{HR}_{\text{rest}} \] (2.7)

2.5.1.1. Heart rate abnormalities

Tachycardia [41]

Tachycardia is a resting heart rate of more than 100 beats per minute. This number can vary as smaller people and children have faster heart rates than adults do.

Bradycardia [41]

Bradycardia is defined as a heart rate of less than 60 beats per minute although it is seldom symptomatic until below 50 bpm. Trained athletes tend to have slow resting heart rates, and resting bradycardia in athletes should not be considered abnormal if the individual has no symptoms associated with it. Again, this number can vary as smaller people and children have faster heart rates than adults do.

Miguel Indurain, a cyclist and five-time Tour de France winner, had a resting heart rate of 28 beats per minute, one of the lowest ever recorded in a healthy human [28].
During our experiments, we used the data in Table 2 as our reference point for defining tachycardia and bradycardia. The algorithm used to calculate the heart rate via the contactless MCG device is presented in Chapter 4, section 4.5.2.1.
CHAPTER 3: TECHNICAL CONCEPTS AND TECHNICAL BACKGROUND

3.1. Introduction

The contactless MCG device uses a 2.45 GHz radar sensor (KMY24), which works in continuous waves, based on the Doppler effect. Therefore, this chapter reviews Doppler effect theory as well as radar theory.

3.1. Doppler effect

The Doppler effect, named after Christian Doppler, is the change in frequency and wavelength of a wave as perceived by an observer moving relative to the source of the waves. For waves that propagate in a wave medium, such as sound waves, the velocity of the observer and of the source are considered relative to the medium in which the waves are transmitted. The Doppler effect may, therefore, result from either motion of the source or motion of the observer. Each of these effects is analyzed separately. For waves that do not require a medium, such as light or gravity in special relativity, only the relative difference in velocity between the observer and the source needs to be considered [49]. The received frequency is described as:

\[ f_{\text{received}} = \frac{f_{\text{source}} \pm v}{c} \]
\[ \omega_r = \omega_o \left(1 + \frac{v}{c} \cos \alpha \right) \] (3.1)

Where:

\( \omega_r \): Corresponds to the received frequency

\( \omega_o \): Corresponds to the frequency of the transmitted wave

\( v \): Corresponds to the relative speed

\( c \): Corresponds to the propagation speed of the wave

\( \alpha \): Corresponds to the angle

Several applications of the Doppler effect are known. Of these, the most commonly experienced is the acoustic effect. The siren of a passing ambulance vehicle seems to have a higher pitch when approaching and a lower pitch when receding. In acoustics, the effect is more complicated than in electromagnetism, since the sound wave propagates using a medium (generally the air). Hence, not only the relative movement between transmitter and receiver, but also the movement between each of them and the medium has to be taken into account. It is necessary to distinguish between moving sender and moving receiver.
The acoustic effect was used as the first experimental proof of the Doppler theory, and the experiment was performed in 1845 by the Dutch physicist Christoph Buys Ballot. A group of trumpet players was placed on a moving train and instructed to play a defined note. Another group was placed next to the tracks and was supposed to play the note they perceived, which turned out to be at a different frequency, in accordance with Doppler's equations [6, 9].

The acoustic Doppler effect is also used in applications of ultrasound, particularly in medicine. Motions of heart valves or blood flow in vessels can be visualized. All velocities on the acoustic axis are displayed together (in Continuous Wave US) or local events in a selected region of interest are shown (in Pulsed Wave US). Different velocities are visualized in different colours. Several flow phenomena can be calculated from the flow velocity and its changes during the heart cycle [10].

If the receiver and the source are approaching each other then the received frequency $\omega_r$ is higher than the base frequency, $\omega_0$, and if the receiver and source are departing from each other, then the received frequency is lower than the base frequency. The speed is defined as positive for a decreasing distance, and as negative for an increasing distance. The Doppler effect doubles during a radar measurement as the beam passes twice through the distance from radar to target. Considering
a direct movement with the angle of $\alpha=0$, $\cos \alpha=1$, the Doppler equation simplifies to:

$$\omega_r = \omega_0 + \omega_D = \omega_0 \left(1 + \frac{\omega}{c}\right)$$  \hspace{1cm} (3.2)

The Doppler effect also occurs with light. It is, for example, the source of the so-called red-shift and blue-shift of moving stars and galaxies.

3.2. Radar

Radar is generally understood to mean a method by means of which short electromagnetic waves are used to detect distant objects and determine their location and movement. The term RADAR is an acronym for ‘Radio Detection And Ranging’ [41].

A complete radar measuring system is comprised of a transmitter with an antenna, a transmission path, a reflecting target, a further transmission path (usually identical to the first one), and a receiver with an antenna. Two separate antennas may be used, but often just one is used for both transmitting and receiving the radar signal. A well-known use of the Doppler effect is the measurement of speed with radar; for example, for the surveillance of speed limits.

To characterize electromagnetic waves, the relevant factors in addition to intensity are their frequency, $f$, and wavelength, $\lambda$, which are
linked by way of their propagation rate, c. The following interrelationships exist:

\[ \begin{align*}
    c &= \lambda f \\
    \lambda &= \frac{c}{f} \\
    f &= \frac{c}{\lambda}
\end{align*} \quad (3.3) \]

The propagation rate \( c \) is equal to the velocity of light. In a vacuum it amounts to exactly \( c_0 = 299 792 458 \text{ m/s} \), or approx. \( 3 \times 10^8 \text{ m/s} \). In gases, it is only negligibly lower. Technical characteristics of a radar system and its range for a monostatic radar can be described as:

\[ P_r = \frac{P_s G^2 \sigma \lambda^2}{(4\pi)^3 R^4 L} \quad (3.4) \]

As is shown in the formula, the power \( P_r \) is reciprocally proportional to the forth power of the distance \( R \), since \( R \) is passed through twice; this is called the \( R^4 \)-Law. In these equations, \( P_s \) is the transmitting power, \( G \) is the gain and \( \lambda \) is the wavelength of the antenna. \( P_s, \sigma \) and \( \lambda \), show the technical characteristics of the transceiving system. \( R \) is the distance and \( \sigma \) is the “radar cross section”. Together, these are the characteristics of the target. Losses of the transceiver are \( L_1 \), losses of signal processing are \( L_2 \), and losses due to target characteristics are \( L_3 \):

\[ L = L_1 + L_2 + L_3 \quad (3.5) \]
The most important parameter for the design of a radar system is the maximum range, $R_{\text{max}}$. Given the minimal detectable power, $P_{\text{MDS}}$, $R_{\text{max}}$ results for a target with the known radar cross section $\sigma$ are expressed as:

$$R_{\text{max}} = \sqrt[4]{\frac{P_s}{P_{\text{MDS}}}} \frac{G^2 \sigma \lambda^2}{(4\pi)^3 L}$$  \hfill (3.6)

### 3.2.1. Transit time [41]

A way to measure the distance to an object is to transmit a short pulse of radio signal (electromagnetic radiation), and measure the time it takes for the reflection to return. The distance is one-half the product of the round-trip time and the speed of the signal (because the signal has to travel to the target and then back to the receiver). Since radio waves travel at the speed of light (186,000 miles per second or 300,000,000 meters per second), accurate distance measurement requires high-performance electronics.

In most cases, the receiver does not detect the returned signal while the signal is being transmitted. Through the use of a device called a duplexer, the radar switches between transmitting and receiving at a predetermined rate. The minimum range is calculated by measuring the
length of the pulse multiplied by the speed of light, divided by two. In order to detect closer targets one must use a shorter pulse length.

A similar effect imposes a maximum range as well. If the return from the target comes in when the next pulse is being sent out, once again the receiver cannot tell the difference. In order to maximize range, one wants to use longer times between pulses, commonly referred to as pulse repetition times (PRT).

These two effects tend to be at odds with each other, and it is not easy to combine both good short-range performance and good long-range performance in single radar. This is because the short pulses needed for a good minimum range broadcast have less total energy, making the returns much smaller and the target harder to detect. This can be offset by using more pulses, but this shortens the maximum range, in turn. Therefore, each radar uses a particular type of signal. Long-range radars tend to use long pulses with long delays between them, and short-range radars use smaller pulses with less time between them. This pattern of pulses and pauses is known as the pulse repetition frequency (or PRF), and is one of the main ways to characterize a radar. As electronics have improved, many types of radar now can change their PRF, thereby changing their range. The newest radars actually fire two pulses during one cell; one for short range, approximately 9.6 km, and another signal for longer ranges (~96.6 km).
The distance resolution and the characteristics of the received signal, as compared to noise, depend heavily on the shape of the pulse. The pulse is often modulated to achieve better performance, thanks to a technique known as pulse compression.

3.2.2. Frequency Modulation

Another method of measuring distance by radar is based on frequency modulation. Frequency comparison between two signals is considerably more accurate, even with older electronics, than just the time of a signal. By changing the frequency of the returned signal and comparing it with the original, the difference can be easily measured. This technique can be used in continuous wave radar, and is often found in aircraft radar altimeters. In these systems, a "carrier" radar signal is frequency modulated in a predictable way, typically varying up and down with a sine wave or saw-tooth pattern at audio frequencies. The signal is then sent out from one antenna and received on another, typically located on the bottom of the aircraft.

Since the signal frequency changes, by the time the signal returns to the aircraft the broadcast has shifted to some other frequency. The amount of that shift is greater over longer times, so greater frequency differences mean a longer distance, the exact amount being the "ramp speed" selected by the electronics. The amount of shift is therefore directly related to the distance traveled, and can be displayed on an instrument. This signal processing is similar to that used in speed-detecting Doppler
radar. Examples of systems using this approach include AZUSA, MISTRAM, and UDOP [41].

3.2.3. Speed measurement

Speed is the change in distance of an object with respect to time. Thus, the Doppler-based system for measuring distance, combined with a memory capacity able to locate the previous target, is enough to measure speed. If the transmitter's output is coherent (phase synchronized), there is another effect that can be used to make almost instant speed measurements, which is known, once again, as the Doppler effect. Most modern radar systems use this principle in the pulse-Doppler system. Return signals from targets are shifted away from this base frequency via the Doppler effect, enabling the calculation of the speed of the object relative to the radar. The Doppler effect is only able to determine the relative speed of the target along the line of sight from the radar to the target. Any component of target velocity perpendicular to the line of sight cannot be determined by using the Doppler effect alone, but it can be determined by tracking the target's azimuth over time.

It is also possible to make a radar without any pulsing, known as a continuous-wave radar (CW radar), by sending out a very pure signal of a known frequency. CW radar is ideal for determining the radial component of a target's velocity, but it cannot determine the target's range. CW radar is typically used by traffic enforcement to measure vehicle speed quickly.
and accurately where range is not important. This is the type of radar we have used in the research that we outline in this thesis.

3.2.4. Frequency bands

The traditional band names (Table 3) originated as code-names during World War II and are still in military and aviation use throughout the world in the 21st century. They have been adopted in the United States by the IEEE (The Institute of Electrical and Electronics Engineers), and internationally by the ITU (International Telecommunication Union). Most countries have additional regulations to control which parts of each band are available for civilian or military use.

Other users of the radio spectrum, such as the broadcasting and electronic countermeasures (ECM) industries, have replaced the traditional military designations with their own systems.

Table 3 shows the radar frequency bands. The frequency used in the mechanocardiography device is in the range of S-Band (2.45 GHz).

Table 3. Radar frequency bands [39]

<table>
<thead>
<tr>
<th>Band Name</th>
<th>Frequency Range</th>
<th>Wavelength Range</th>
<th>Notes</th>
</tr>
</thead>
<tbody>
<tr>
<td>HF</td>
<td>3–30 MHz</td>
<td>10–100 m</td>
<td>Coastal radar systems, over-the-horizon radar (OTH) radars; 'high frequency'</td>
</tr>
<tr>
<td>P</td>
<td>&lt; 300 MHz</td>
<td>1 m+</td>
<td>'P' for 'previous', applied retrospectively to early</td>
</tr>
<tr>
<td>Band</td>
<td>Frequency Range</td>
<td>Typical Wave Length</td>
<td>Application Notes</td>
</tr>
<tr>
<td>------</td>
<td>-----------------</td>
<td>---------------------</td>
<td>--------------------</td>
</tr>
<tr>
<td>VHF</td>
<td>50–330 MHz</td>
<td>0.9-6 m</td>
<td>Very long range, ground penetrating; 'very high frequency'</td>
</tr>
<tr>
<td>UHF</td>
<td>300–1000 MHz</td>
<td>0.3-1 m</td>
<td>Very long range (e.g. ballistic missile early warning), ground penetrating, foliage penetrating; 'ultra high frequency'</td>
</tr>
<tr>
<td>L</td>
<td>1–2 GHz</td>
<td>15–30 cm</td>
<td>Long range air traffic control and surveillance; 'L' for 'long'</td>
</tr>
<tr>
<td>S</td>
<td>2–4 GHz</td>
<td>7.5–15 cm</td>
<td>Terminal air traffic control, long-range weather, marine radar; 'S' for 'short'</td>
</tr>
<tr>
<td>C</td>
<td>4–8 GHz</td>
<td>3.75-7.5 cm</td>
<td>Satellite transponders; a compromise (hence 'C') between X and S bands; weather</td>
</tr>
<tr>
<td>X</td>
<td>8–12 GHz</td>
<td>2.5-3.75 cm</td>
<td>Missile guidance, marine radar, weather, medium-resolution mapping and ground surveillance; in the USA the narrow range 10.525 GHz ±25 MHz is used for airport radar. Named X band because the frequency was a secret during WW2.</td>
</tr>
<tr>
<td>K_u</td>
<td>12–18 GHz</td>
<td>1.67-2.5 cm</td>
<td>High-resolution mapping, satellite altimetry; frequency just under K band (hence 'u')</td>
</tr>
<tr>
<td>K</td>
<td>18–27 GHz</td>
<td>1.11-1.67 cm</td>
<td>From German <em>kurz</em>, meaning 'short'; limited use due to absorption by water vapour, so K_u and K were used instead for surveillance. K-band is used for detecting clouds by meteorologists, and by police for detecting speeding motorists. K-band radar guns operate at 24.150 ± 0.100 GHz.</td>
</tr>
<tr>
<td>K_a</td>
<td>27–40 GHz</td>
<td>0.75-1.11 cm</td>
<td>Mapping, short range, airport surveillance; frequency just above K band (hence 'a') Photo radar, used to trigger cameras which take pictures of license plates of cars running red lights, operates at 34.300 ± 0.100 GHz.</td>
</tr>
<tr>
<td>mm</td>
<td>40–300 GHz</td>
<td>7.5 mm - 1 mm</td>
<td>Millimetre band, subdivided as below. The letter designators appear to be random, and the frequency ranges dependent on waveguide size. Different groups assign multiple letters to these bands. These</td>
</tr>
</tbody>
</table>
are from Baytron, a now defunct company that made test equipment.

<p>| | | |</p>
<table>
<thead>
<tr>
<th></th>
<th></th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td>Q</td>
<td>40–60 GHz</td>
<td>7.5 mm - 5 mm</td>
</tr>
<tr>
<td>V</td>
<td>50–75 GHz</td>
<td>6.0–4 mm</td>
</tr>
<tr>
<td>W</td>
<td>75–110 GHz</td>
<td>2.7 - 4.0 mm</td>
</tr>
</tbody>
</table>
CHAPTER 4: MECHANOCARDIOGRAPHY (MCG) USING RADAR PRINCIPLES

4.1. Introduction

In this chapter, we present the hardware involved in radar mechanocardiography (MCG). The safety of the device is a main concern, so in section 4.1 we provide a review of the safety for humans of the frequency it uses. The new MCG device design is shown in section 4.2. Since the signal coming from the subject’s monitored organ is very weak, there is a need for good filtering and amplification, and we describe this fully with regard to the signal’s simulation under the “filtering and amplification” section. In the said section, we describe some measurements that were taken to fully understand the components of the sensor used in this device. The device consists of a transmitter, a receiver, a band-pass filter, and an amplifier.

The output from the filter and the amplifier is sent to the microprocessor for further processing. The microprocessor used in this design is an 8 bit Atmega128 with 32 general-purpose working registers. In section 4.5, we describe the microprocessor. Subsequently, in the “ADC” section, we discuss analogue to digital conversion.
In section 4.7, we describe the device's potential for connection to a computer. In section 4.8, we outline the connection of the microprocessor to the TFT display (thin film transistor liquid crystal display).

The device has a unique feature, which is its self-contained display, eliminating the need for connecting the device to another system, like a computer. The display is described in section 4.9, following instructions regarding the device’s keypad. Furthermore, it is possible to connect the contactless mechanocardiography device to the MATLAB program for further signal processing. We describe this in section 4.11.

Figure 4.1 shows the PCB board we have implemented in the device. Figure 4.2 shows the principal design of the radar-based MCG device, which contains a 2.45 GHz oscillator, a transmitter and a receiver in the same housing. The transmitter transmits continuous-wave radio frequency energy towards the subject’s body. The returned signal is frequency shifted due to the Doppler effect. This effect permits the measurement of slight body movements, from which the MCG receiver can obtain the heart’s mechanical signal. The output from the receiver needs to be filtered and amplified before being sent to the microprocessor for further processing.

The contactless mechanocardiography device runs on 2 AA batteries. For the sensor we need 12V and for the microprocessor we need 5V, so a voltage regulator has been designed for the device and is fully described under in section 4.6.
The cut-off frequencies for the band-pass filter shown in the schematic are 1 Hz and 100 Hz and the gain of the amplifier is around 800. After filtering and amplifying the MCG signal, it is sent to the A/D unit and then to the ATMEL CPU for further processing. The CPU is connected to the thin-film transistor (TFT) display via its SPI (Serial Peripheral interface) port, so the data from the microprocessor then becomes visible on-screen.

Figure 4.1. Schematic diagram of the mechanocardiography device
4.2. Operating frequency of the device and safety issues

The frequency range used by this device is 2.45 GHz, which is a safe frequency for the human body, as shown in a number of studies. This is the same frequency range as used in cellular phones. A main reason for choosing this frequency for the device is that under current telecommunications regulations, only motion sensors operating on the 2.45 GHz and 24.125 GHz frequency bands can be marketed worldwide. The sensors working in the frequency range of 24.125 GHz are difficult to implement with surface mount devices (SMDs). For this reason, we have chosen a KMY24 Doppler sensor for the contactless mechanocardiography device. This particular sensor already has German BZT G127520H approval to operate in the 2.45 GHz band. Advantages of the new motion sensor include its suitability for worldwide approval, its low insertion height, and its ability to detect direction.

It is important to note that the modules which contain transceivers also emit RF radiation at the back if no protective shielding is provided. The ratio of front-to-back radiation is about 3 to 1 for the KMY24. The reason for this is the relatively low gain of the transmit/receive antenna. It is often used to operate automatic light switches in building management systems, for example.
4.3. Design of the main sensing device

The Doppler radar used in the design of our radar-based MCG device consists of a four layered epoxy multilayer board with SMD-components. The Doppler radar works in continuous wave mode and contains a 2.45 GHz oscillator transmitter and a receiver in the same housing.

In order to minimize the emission of harmonics in the sensor, we employ triple filter structures. The module also contains an optimized patch antenna with minimized dimensions.

Table 4 lists the parameters of the sensor used in the device. Figure 4.3 shows the typical far-field radiation pattern of the sensor in the
x-z and y-z planes. Radio frequency radiation is emitted from the radar with a front to back ratio of 3 to 1, the reason for which is the relatively low gain of the antenna and the shielding [20].

![Diagram](image)

**Figure 4.3. Vertical/horizontal diagram of the patch antenna used in radar based MCG device at 2.45 GHz [20]**

In the radar sensor, two mixer diodes are used as receivers. The diode nearer to the antenna provides the output voltage \( UD_1 \), and the diode located farther provides the output voltage \( UD_2 \). The mixer diodes are driven by the oscillator with a defined phase difference. According to the data sheet, the phase shift between the two mixer diodes is an odd-numbered multiple.
Table 4. Characteristics of the RF sensor used in Radar MCG device

<table>
<thead>
<tr>
<th>Supply Voltage</th>
<th>$V_s$</th>
<th>10.8</th>
<th>12</th>
<th>15.6</th>
<th>$V$</th>
</tr>
</thead>
<tbody>
<tr>
<td>Operating frequency</td>
<td>$f_0$</td>
<td>2.40</td>
<td>2.45</td>
<td>2.48</td>
<td>GHz</td>
</tr>
<tr>
<td>Operating Current</td>
<td>$I_{op}$</td>
<td>23</td>
<td>$mA$</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Equivalent Isotropically Radiated Power @ $f_0$</td>
<td>EIRP</td>
<td>-</td>
<td>8</td>
<td>-</td>
<td>$dBm$</td>
</tr>
</tbody>
</table>

The phase noise of the oscillator is reduced by using a transistor with a minimal transition frequency of $f_T$. The basic rule for oscillator design is represented by the following formula:

$$f_T = 2 \mu f_0$$  \hspace{1cm} (4.1)

The oscillator used in the sensor design is BFR 92P with a transition frequency of 5 GHz, a ceramic coaxial resonator, several 0805 series resistors and capacitors, and various micro-strip conductors. The oscillator output power is about 10 dBm (Figure 4.4).
Figure 4.4. The oscillator used in the radar MCG device
Figure 4.5. 'a' shows block diagram of the principle design of the KM24 microwave sensor. 'b' and 'c' shows receiving signals VD1 and VD2, in approaching and descending mode.
\[ V_{D1} = A_1 \cos(\omega t + \varphi_1) \cos\left(\omega t \pm \frac{2\nu}{c} + \varphi_1 + 2\varphi_2\right) \]  
\[ V_{D2} = A_2 \cos(\omega t) \cos\left(\omega t \pm \frac{2\nu}{c} \omega t + 2\varphi_1 + 2\varphi_2\right) \]  
\[ \cos a \cos b = \frac{1}{2} \left[ \cos(a - b) + \cos(a + b) \right] \], follows:

\[ V_{D1} = \frac{A_1}{2} \cos\left(\pm \frac{2\nu}{\epsilon} \omega t - 2\varphi_2\right) \]  
\[ V_{D1} = \frac{A_1}{2} \cos\left(\pm \frac{2\nu}{\epsilon} \omega t - 2\varphi_1 - 2\varphi_2\right) \]  
\[ \varphi_1 = \frac{\pi}{4}, \frac{3\pi}{4}, \ldots \]  

4.4. Filtering and amplification

Electrical filters were in use long before IC operational amplifiers (Op-Amps) were available. Typically, they were constructed with passive components such as resistors, capacitors and inductors. This is the reason they are called passive filters. With low-cost and reliable active amplifying devices available, such as Op Amps, filter designs are now mainly active types.

The band-pass filter we have designed consists of low-pass and high-pass filters. A low-pass (LP) filter passes frequencies below the cutoff frequency of 100 Hz and attenuates frequencies above the cutoff frequency. The high-pass (HP=1Hz) does the opposite.

In designing the filter and amplification board, we used SwitchCAD for the simulation, and FilterPro with Multiple Feedback (MFB). There were several options to choose from in designing the filter, including:
Butterworth, Chebychev, Bassel, Linear phase, 0.5-Linear Phase, Custom Fn&Q and Gaussian filters. As it is very important to minimize the ripples in the band-pass, for this design we chose a Butterworth filter, which is as flat as mathematically possible in the pass-band. Butterworth filters are also known as 'maximally flat magnitude' filters.

In FilterPro, the response of the filter is displayed on a graph, showing gain, phase and group delay over frequency. The circuit type that we chose is MFB. Some of the advantages of the MFB configuration are the minimal number of circuit elements, low output resistance, convenience in cascading with other stages, good stability characteristics and low component sensitivity. By using MFB, we preclude common-mode error in the filter.
The cut-off frequency in our contactless mechanocardiography device is 1 Hz to 100 Hz. The cut-off frequency as defined in physics and electrical engineering is different from that defined in the field of "electronics".

In physics and electrical engineering, the term "cut-off frequency", or corner frequency, represents a boundary in the system response, at which point energy entering the system begins to be attenuated or
reflected instead of transmitted. Common examples are the cut-off frequency of an electronic circuit: either the lowest or the highest frequency at which the output of the circuit deviates less than 3 dB from the nominal value. The cut-off frequency can also refer to the plasma frequency, or to some concepts related to renormalization in quantum field theory.

In electronics, cut-off frequency, or corner frequency (fc), is the frequency either above which or below which the power output of a circuit, such as a line, amplifier, or electronic filter, is 1/2 the power of the band-pass. Since voltage, \( V_2 \), is proportional to power, \( P \), \( V \) is \( \sqrt{\frac{1}{2}} \) of the \( V \) in the band-pass. This happens to be close to \(-3\) decibels, and the cut-off frequency is frequently referred to as the \(-3\) dB point.

Figure 4.7 shows the connections between the filter and amplifier to the KMY24’s jack. Pin 4 on the P1 is supplied with the 12V and Pin1 is the ground for the KMY24. The +5V source voltage is needed for the active components and it comes from the voltage regulator designed for the contactless MCG device (see section 4.6).
For the first stage of filter design, we have chosen to use a micro-
power instrumentation amplifier optimized for single supply operation from
2.2V to 36V (LT1789-10). The output from the radar is fed into the
LT1789. The LT1789-10 is laser trimmed for very low input offset voltage
that drifts high CMRR and high PSRR. The output can handle capacitive
loads up to 1000pF in any gain configuration while the inputs are ESD
(Electro-Static Discharge) protected for the human body (10kV). The
quiescent current is 95μA max and the input and output common mode
swing within 110mV of ground. The gain is set with a single external
resistor. The LT1789-10 maximizes the input common mode range and
dynamic output range when an amplification of 10 or greater is required,
allowing precise signal processing where other instrumentation amplifiers
fail to operate. We will now go on to describe the schematics of the filter
and the amplifier in detail.

In the filter design, the mid-point is provided by the following
schematic and is used as a REF for LT1789-10.
Mid-point is the low resistance, filtered reference voltage of $\frac{1}{2} \text{Vcc}$ for U1. LT1112, used to create the mid point, has high large-signal gain, low input bias current, and low input offset voltage and performs very well in this schematic. U4 works as a voltage buffer and C3 is filter capacitance.
Figure 4.9. LT1789, ESD (Electro-Static Discharge) protected for the human body

U1, LT1789, amplifies the difference between V2 and V1. Resistor R1 determines U1 gain.

\[
\text{Gain of U1} = 1 + \frac{200k}{R1} = 1 + \frac{200}{1.5} = 301
\]  
(4.7)

In the input stage, C4 and C5 are input coupling capacitors (DC blocking), and resistors R7 and R8 provide DC ground for inputs of U1. The result is:

\[
\text{f_in} = \frac{1}{2\pi C4 R7} = 0.5\text{Hz (DC blocking)}
\]  
(4.8)
Figure 4.10. Simulation for the output of LT1789-10

Figure 4.11. Op-amp LT6221 used in design of the filter

In our design of the low-pass filter stage, we use LT6221, named U2, which is a single/dual/quad, low power, high speed rail-to-rail input
and output operational amplifier with excellent DC performance. The LT6221 features reduced supply current, lower input offset voltage, lower input bias current and higher DC gain than other devices with comparable bandwidth. A feedback resistor of 255K is used to set up gain for U2, so care must be taken to ensure that the pole formed by the feedback resistors and the total capacitance at the inverting input do not degrade stability.

The LT6221’s input stage is also protected against a large differential input voltage of 1.4V or higher by a pair of back-to-back diodes, D5/D8, to prevent the emitter-base breakdown of the input transistors. The current in these diodes should be limited to less than 10mA when they are active.

In the output stage, we have R4, R5, R6, C1, C2, and U2 as a 2nd order 125 Hz low pass filter with a gain of:

\[
R_6/R_5 = \frac{255}{47} = 5.42
\]  

(4.9)

By experiment, the following result for the low-pass filter is achieved:
In designing a filter, the number of poles affects the filter's rolloff rate. Here, the filter has 3 poles. U3 is a dual low power precision, picoamp input operation amplifier, which works as the voltage buffer.

\[ V = \frac{R10}{(R9+R10)} \cdot Vcc \]  \hspace{1cm} (4.10)
Figure 4.13. LT1112 used to design the filter in MCG

Frequency range is at about 0.5 – 125 Hz

Gain = 301 * 5.42 = 1631  \hspace{1cm} (4.11)

Figure 4.14. Simulation for figure 4.13
Figure 4.15. The filtering and the amplification schematic

Figure 4.16. Amplifier and filter on the PCB
Figure 4.17. The frequency response of the system

It should be noted that the signal coming from the heart is very weak, so we need very high gain amplification. The filter designed for this purpose works very well in conjunction with cancelling-out of the breathing signal by using a high pass filter of 1 Hz and a very ideal low pass filter of 100 Hz. The band pass filter has minimal ripples on it due to the Butterworth filter design approach. The distributed gain design is used in order to optimally minimize the unstable gain.
4.5. Microprocessor

The output from the filter and the amplifier is the amplified, filtered-out signal of the radar that comes from the heart and is sent to the microprocessor for further processing. The microprocessor used in this design is an 8 bit Atmega128 with 32 general-purpose working registers. The ATmega128 is a low-power CMOS 8-bit microcontroller based on the AVR enhanced RISC architecture. A block diagram of the Atmega128 is shown in figure 4.18.
Figure 4.18. Block diagram of Atmega128
All the 32 registers are directly connected to the Arithmetic Logic Unit (ALU), allowing two independent registers to be accessed in one single instruction, executed in one clock cycle. The resulting architecture is more code-efficient while achieving throughputs up to ten times faster than conventional CISC microcontrollers.

The Atmega128 provides the following: 128K bytes of in-system, programmable flash with read-while-write capabilities, 4K bytes EEPROM, 4K bytes SRAM, 53 general purpose I/O lines, 32 general purpose working registers, a Real Time Counter (RTC), four flexible timer/counters with compare modes and PWM, 2 USARTs, a byte oriented two-wire serial interface, an 8-channel, 10-bit ADC with optional differential input stage with programmable gain, a programmable watchdog timer with internal oscillator, an SPI serial port, IEEE std. 1149.1 compliant JTAG test interface, also used for accessing the on-chip debug system and programming, and six software-selectable power saving modes.

The idle mode stops the CPU while allowing the SRAM, time/counters, SPI port, and interrupt system to continue functioning. The power-down mode saves the register contents but freezes the oscillator, disabling all other chip functions until the next interrupt or hardware reset. In power-save mode, the asynchronous timer continues to run, allowing the user to maintain a timer base while the rest of the device is sleeping. The ADC noise reduction mode stops the CPU and all I/O modules except the asynchronous timer and ADC, to minimize switching noise during ADC
conversions. Figure 4.19 shows the microprocessor’s connections on the board.

![Figure 4.19. Atmega connections on the board](image)

The oscillator’s frequency is 14.7456 MHz. In standby mode, the crystal/resonator oscillator runs while the rest of the device is sleeping. This allows very fast start-up combined with low power consumption. In extended standby mode, both the main oscillator and the asynchronous timer continue to run. The timing is based on interrupt from 8-bit Timer0 overflow. In the interrupt Timer0 must be preloaded with the value 240 (0xf0). The fastest interrupt frequency is as follows:

\[
F = \frac{14.7456 \text{MHz}}{1024 \times 240} = 60 \text{ IRQ/sec}
\]  

(4.12)

Which means it happens every 16.66 ms.
4.5.1. ADC conversion

Based on the fastest interrupt frequency of $F = 60 \text{IRQ/sec}$, the display fills up for $t = 0.01666 \times 176 = 2.9 \text{sec}$, which is a very good range, showing about three beats. If more beats need to be displayed, it is possible to show them faster by reloading the Timer0 inside the IRQ (interrupted request) with a value smaller than 0xf0, keeping the same interrupt rate that we are averaging. For example, to display six heartbeats for 5.8 sec, then for every displayed value, we take the average of two sequential measurements. Before sending the value to the display, it needs to be right shifted twice (divided by 4) to fill up the screen in a vertical direction (132 pixels).

Port F serves as the analog input to the A/D converter. AVCC is the supply voltage pin for Port F and the A/D converter. AREF is the analog reference voltage pin for the A/D converter.
Figure 4.20. Modification on Atmega128 module, AVCC is filtered by L and C

According to the Atmega128 data sheet, for all unused ports the pull-up resistors must be enabled. The reference (Ref) voltage for the ADC is connected to VCC (5V) and a filter capacitor is installed. ADC0 is connected to the radar output and ADC2 to the battery voltage.

The algorithm for the ADC works in such a way that a counter with three states is incremented on every Time0-interrupt. At each stage, the next Mux-channel is selected and an ADC conversion started. Therefore, during every time period, three measurements are taken, which are attributed to three different variables.

The result from the ADC in its original format is also stored in a cycle buffer with a length L=128 bits, where the function “beat_detect” calculates the heart beat rate in beats/sec. After the buffer is filled up, the
two biggest values are taken and the peaks are subtracted. The result is a
number of ticks 16.66ms/each, so the beat period is \( T[\text{sec}] = \text{ticks} \times 0.0166 \) and the beat rate is \( BR = \frac{1}{T} \).

4.5.2. Programming of the microprocessor

To program the Atmel micro controller we used WinAVR software in
\( C \) (WinAVR is a good open source C compiler). In principle, the output of
data to the display always functions in the same way:
1) Communication with the display is carried out by setting the "window co-ordinates", within which the display's data (pixels) range has to be entered.
2) All sent data is interpreted by the display as a pixel and then set. It starts with the upper left corner of the opened "window" and moves forward one position to the right automatically. If the right border of the window is reached, the pointer automatically moves to the beginning of the next line.
3) The wire DC (data/communication ID) of the display tells it if the received data contains pixel information or a command. "1" on the wired DC signals a command, a "0" means pixel.
4) A pixel always needs 2 bytes, as each pixel can consist of one of 65536 colours. For a full-page picture, the software has to transmit 132x176pixelsx2 bytes, which is 46 Kbytes of data, or 371712 bits.
4.5.2.1. C program for calculating the heartbeat

We employed the following code in programming the microcontroller and in order to calculate the heartbeat. All the processing is done at this stage, so there is no need for external computers.

"LCD_Print(text, 50, 80, 1, 1, 1, blue, white);"
"sprintf(text, "--- bpm");"
while(1)
{
if (tmr0 > 0)
{
    tmr0 = 0;
    putchar(newADC);
}

    The ADC conversion results are read and sent to the UART (Universal Asynchronous Receiver/Transmitter), then the data is shifted 3 times to the right in order to use the 7 most significant bits to fit the data vertically into the display. The data is recorded in a buffer to facilitate a moving average calculation:

data = read_adc(0);
sampleTime = time;
// Send the data over the UART
outputData(data);
newADC = (data & 0x3FF) >> 3;
mvAvg[mvHead++] = data >> 2;

The following code shifts the graphic leftwards, if the scope line reaches the right corner of the LCD:
if (mvTail < MOVING_WINDOW_SIZE)
{
    mvTail++;
}
else
{
    mvTail = MOVING_WINDOW_SIZE;
}
if (mvHead >= MOVING_WINDOW_SIZE)
{
    mvHead = 0;
}

The moving average is calculated. The user then calculates the peaks by counting how many times the signal goes over and then under the horizontal cursor line, which is adjustable.

"avg = mvCalc(mvAvg, mvTail) >> 1;"
if (below && (avg > peak))
{
    below = 0;
}
else if (!below && (avg < peak))
{
    below = 1;
    peakTimes[peakHead++] = sampleTime;
    if (peakHead >= PULSES)
    {
        peakHead = 0;
    }
}
if (peakTail < PULSES)
{
    peakTail++;
}
if (peakTail >= PULSES)
{

    The program checks to see if it has collected enough peaks to calculate bpm and then assigns the bpm value a text variable so it can be displayed later on. It then checks if it has detected any pulses in a reasonable amount of time. If the bpm is lower than 30, then the bpm count is reset. The program displays the data only if the freeze button is not pressed and deletes old pixel data before displaying new ones. In addition, it displays a short line on the left that represents the cursor, and if the signal goes over the cursor, a peak/beat is counted.

    newestTime = sampleTime;
    oldestTime = peakTimes[peakHead];
    timeDiff = newestTime - oldest Time;
    "sprintf(text, "%03d bpm", (60*000 / 16) / (timeDiff / (PULSES-1)) );
    peakTail = PULSES;

}

#define NO_PULSE_LIMIT ((1000/16) * 4)
if (((time - peakTimes[0]) > NO_PULSE_LIMIT)
{
    peakHead = 0;
    peakTail = 0;
    time = 0;
    peakTimes[0] = 0;
    "sprintf(text, "--- bpm"),";
}
if (step >= numSteps)
if (!freeze)
{
"LCD_Draw(x+1, 0, x+1, 132, white);
"LCD_Draw(x+2, 0, x+2, 7, blue);
"LCD_Draw(0, 131-peak, 7, 131-peak, blue);"

It then displays the averaged data:

"LCD_Plot(x, 131-(avg), green);"

followed by the actual data:

"LCD_Draw(x, 131-prevADC, x+1,131-newADC, red);" // the red line shows the heartbeat in realtime
prevADC = newADC;
x++;
if (x >= 175)
x = 0;
}
step = 0;
}
else
{
step++;
}
If the freeze button is not pressed then it displays the beats per minute:

```c
if (! freeze)
{
"LCD_Print(text, 108, 118, 1, 1, 1, blu"e", white);"
}
```

4.6. **Switching voltage design used in the contactless MCG**

The contactless mechanocardiography system is powered by 2 AA batteries or by connecting the device to the USB (Universal Serial Bus) port of a computer. The nominal output voltage of single-use AA batteries is 1.5 volts, while NiCd (Nickel Cadmium) rechargeable batteries have a nominal voltage of 1.2 V.

![Figure 4.21. The block diagram of the DC/DC convertor](image)
AA batteries are used to supply the needed voltage for the oscillator and the microprocessor. The oscillator operates on 12V and the microprocessor needs 5V. For that reason, we designed a voltage converter circuit using an LT1944. The LT1944 is a dual micro-power step-up DC/DC converter in a 10-pin MSOP package. Each converter is designed with a 350mA current limit and an input voltage range of 1.2V to 15V. Both converters feature a quiescent current of only 20μA at no load, which further reduces to 0.5μA in shutdown. A current limited, fixed off-time control scheme conserves operating current, resulting in high efficiency over a broad range of load current. Figure 4.22 shows the block diagram of the switching voltage converter.

![Block diagram of the voltage converter](image)

Figure 4.22. Block diagram of the voltage converter

The LT1944 uses a constant off-time control scheme to provide high efficiencies over a wide range of output current. Its operation can be best understood by referring to the block diagram in Figure 4.23. Q1 and
Q2 along with R3 and R4 form a band gap reference used to regulate the output voltage. When the voltage at the FB1 pin is slightly above 1.23V, comparator A1 disables most of the internal circuitry. Output current is then provided by capacitor C2, which slowly discharges until the voltage at the FB1 pin drops below the lower hysteresis point of A1 (typical hysteresis at the FB pin is 8mV). A1 then enables the internal circuitry, turns on power switch Q3, and the current in inductor L1 begins ramping up.

Once the switch current reaches 350mA, comparator A2 resets the one shot, which turns off Q3 for 400ns. L1 then delivers current to the output through diode D1 as the inductor current ramps down. Q3 turns on again and the inductor current ramps back up to 350mA, then A2 resets the one shot, again allowing L1 to deliver current to the output. This switching action continues until the output voltage is charged up (until the FB1 pin reaches 1.23V), then A1 turns off the internal circuitry and the cycle repeats. The LT1944 contains additional circuitry to provide protection during start-up and under short-circuit conditions. When the FB1 pin voltage is less than approximately 600mV, the switch-off time is increased to 1.5 ms and the current limit is reduced to around 250mA (70% of its normal value). This reduces the average inductor current and helps minimize the power dissipation in the power switch and in the external inductor and diode. The second switching regulator operates in
the same manner. Figure 4.23 shows the DC/DC converter’s circuit design.

![Block diagram of the voltage converter](image)

**Figure 4.23. Block diagram of the voltage converter**

Low ESR (Equivalent Series Resistance) capacitors are used at the output to minimize the output ripple voltage. Multilayer ceramic capacitors are the best choice, as they have a very low ESR and are available in very small packages.

For the input decoupling, a 4.7\( \mu \)F capacitor was used and it was placed close to the LT1944. MBS2040 has a low forward voltage drop and fast switching speed. As a result, it is the best match for the LT1944.

The LT1944 provides energy to the load in bursts by ramping up the inductor current, then delivering that current to the load. If too large of an inductor value or too small of a capacitor value is used, the output ripple voltage will increase because the capacitor will be slightly overcharged in each burst cycle. A 4.7pF feed-forward capacitor in the
feedback network of the LT1944 is used to reduce the output ripple, so by using the 4.7pF capacitor the output voltage ripple is greatly reduced.

As mentioned above, the contactless mechanocardiography device can also be powered via USB. Figure 4.24 shows the schematic of the design. Two MBRS2040 diodes have been used here. Every time the system connects to the USB, D1 starts working and D2 prevents the power from going to the batteries so they do not overcharge or explode.

![Figure 4.24. The schematic of the design](image)

4.7. Interfacing the MCG device to the computer

The results from the ADC output of the microprocessor go to a recoding function that makes a packet ready to be sent through UART to USB and to the host PC. The ADC result takes 2 bytes as the ADC is 10 bits. The physical layer protocol is as follows. The 2 bytes of unsigned hex data from the ADC is converted into ASCII format. The measurements are separated by commas. This format is known as CSV. For example:
A serial-to-USB converter chip FTDI, FT232RL, enables the data to be sent to the computer with an auto baud rate setting.

![FT232R schematic](image)

**Figure 4.24. USB host PC connection, based on FTDI chip FT232R schematic**

4.8. **Connection of the microprocessor to the display**

The connection between the microcontroller and the graphics TFT is done serially. This means the needed data is moved to TFT bit by bit. The display uses a command to reduce the pixel output to a specified area and allows the selection of an output direction. This is very helpful, as it reduces the number of pixels we need to transfer in order to change a
character with a size of 5x8 to 40 pixels. We just open a virtual window of a size of 5x8 pixels and we then send 40-pixel data (which is the individual pixel colour). Due to the extra display memory, we do not have to devise a way to refresh the whole display 60 times a second. If we had to do this, the microcontroller would definitely not be able to do anything else. In our setup, the microcontroller only needs to act if we change something on the screen.

As mentioned above, the data is sent serially to the display. To save our resources and to speed up this serial transmission we use the SPI hardware module built into the Atmega. To do this, we connect the display to the SPI port of the microcontroller and we set data and clock lines. However, the display needs select (CS), reset, and data/command (DC) lines which, if necessary, the software must toggle manually. The display is connected via 5 wires:

a) CS (select)

b) Clock (clock signal)

c) Data (the data bits)

d) DC (data/command identification)

e) Reset (full reset)

Figure 4.25 shows the connection between the microcontroller and the display. Between the microcontroller and display, we use a buffer chip.
Figure 4.25. Schematics of the connection of the display to the microcontroller
4.9. The output window

After determining the output data, an output window is "opened" and the needed pixel data are sent. A character in font displayed in the output window is always exactly 6 pixels wide as well as 8 pixels high. The definition of the output window is determined by the sub-routine LCD-window using four values: the X and Y position of the left upper corner as well as X and Y of the right lower corner.

Using font I with 1x scaling, the lower right corner is always 6 pixels further to the right and 8 pixels down from the top left corner. The distance grows by using larger fonts or by using a scaling factor. The pixels are
simply multiplied by the scaling factor. A sign double in size could of course be defined in data (array) lines at a higher resolution, but this font would then need 4 times the memory of the smaller font (2 times the pixel vertically x 2 times the pixel horizontally). A three-times-larger font would use 9 times the memory. The output direction of the sent bits (pixels) is left to right.

Figure 4.27. Schematic of the connections between Atmega128 and LCD
4.10. Keypad and manual commands

Here we describe the buttons used to operate the contactless mechanocardiography device. Operating the device is very simple and there is no need for the user to have any knowledge of computing. All the instructions are written next to each button. They are as follows:

D0 → Decrease gain; sends a command to a digital pot to step up the resistance. D0 is replaced with G0 because port D0 is needed for I²C (cut trace and place wire to G0 as outlined in the Display3000 manual.)
D1 → Increase speed (more measurements per second)
D4 → Stop play (freeze the screen)
D5 → Increase gain (send a command to a digital pot for lowering the resistance)
D6 → Decrease speed
D7 → Play (release the data plot)
4.11. Real-time communication with MATLAB [47]

In order to locate, configure, and control serial devices and instruments, we use TMTool, a graphical user interface provided by Instrument Control Toolbox. We can explore available hardware assets using a hierarchical tree.
Communicating with the MCG device to acquire data requires establishing a connection to its communication interface. With TMTool, we can configure, control, and acquire data from the contactless MCG device through a well-established serial interface (Figure 4.30).

![Figure 4.30. TMTool interface](image)

After connecting the contactless MCG device to COM4, we select the ‘Configure’ tab for COM4 and set it to match the serial port parameters. NMEA defines these parameters as baud rate 115200 bits per second, 8 data bits, 1 stop bit and no parity.
After configuring the serial port parameters, we open the connection via the serial port. The connection status changes to ‘Connected’, indicating that communication with the contactless MCG device through the ‘Communicate’ tab is now possible. The NMEA standard specifies that data must be transmitted as ASCII characters in new line-terminated strings. Using the dropdown menu, we set the ‘Receiving data’ parameters to comply with this specification.
We can now send data to the hardware and read the responses. For our example, all we need to do is read the data broadcast from the MCG device. We can confirm that we are communicating properly and that the data of interest are available by clicking the ‘Read’ button several times to observe that properly formatted strings are being acquired in the sequential log. After a few attempts, the MCG device coordinates of interest appear in the results, together with several other NMEA data types.

TMTool automatically updates the MATLAB script generated in the ‘Session Log’ tab. Once we have exported the updated script MATLAB_MCG.m, we can insert MATLAB routines to filter the data stream.

4.11.1. Creating a generic driver

We can incorporate the MATLAB script into a reusable driver by using MIDEEdit, a driver development tool in Instrument Control Toolbox.
MIDEdit lets you incorporate lower-level commands into higher-level commands that are easier to access. After launching MIDEdit from the command line, we create a new generic instrument driver through the File->New context menu.

To configure the MCG device, we browse to the automatically generated MATLAB script in the TMTool session log. This script sets the baud rate and then opens the port. We can now copy this information into a driver. Within MIDEdit, we select the 'Connect' tab from the 'Initialization and Cleanup' node and set the 'Function Style' to M-Code. We then paste the script from TMTool into the function created in the driver editor. Noting that the serial interface may be obtained from the variable `obj` passed into this function, the MATLAB script is adjusted to use the serial port interface. Appendix 3 shows the M-file used to display the contactless MCG signal on a real-time basis on the MATLAB program.
CHAPTER 5: EXPERIMENTAL RESULTS

In this chapter, we outline the results taken from the contactless MCG device experiments we conducted at the CIBER lab, Burnaby Hospital and at SFU’s Kinesiology Department.

At Burnaby Hospital, we investigated the signal received from the device synchronized with the standard known BCG signal. At the kinesiology department, we investigated the signal received from the major blood vessels using the contactless MCG device and compared the data to the signal received from an ultrasound device.

It is our belief that the results discussed below show a significant potential value for our device, especially considering that it is ‘contactless’, and, unlike ultrasound, does not require gel to be applied to the area being monitored. From these initial results, it therefore seems our device could eventually perform well in today’s market for low-cost, effective biomedical technology.

5.1. Heart rate and breathing rate detection

The data from eight subjects were acquired at Burnaby General Hospital. The processed MCG signal was superimposed on the phonocardiograph signal and it was noted that the S1 and S2 sounds of the phonocardiograph signal corresponded to the similar complexes on
the radar MCG signal, showing the correlation of the MCG signal to the heart cycle's events. The contactless MCG methodology can be seen in Figure 5.1.

![Radar-based mechanocardiograph (MCG) signal processing methodology](image)

**Figure 5.1.** Radar-based mechanocardiograph (MCG) signal processing methodology

For detection of breathing rate, the radar signal was low pass filtered under 0.4 Hz and the peaks were counted and compared to the results acquired from a strain gauge transducer that measures the changes in thoracic circumference, using a belt which is fastened to the subject's thorax. As can be seen in Table 5, the accuracy of respiration rate measurement was 91.35 percent across all subjects.

Heart rate was also measured using the radar-based MCG device and was compared to the heart rate calculated from a simultaneous ECG signal for six subjects. The average of the heart rate accuracy on these subjects was calculated to be 92.9 percent. The results of heart rate and breathing rate measurements for each individual subject can be seen in Table 5.
Figure 5.2. Two cycles of synchronous radar MCG, phonocardiograph and ECG signal showing the correlation of cardiac cycle events to the radar MCG signal. Systolic and diastolic complexes can be identified in the radar MCG signal corresponding to S1 and S2 of heart's sounds.

5.2. The mechanocardiography signal synchronized with BCG

The first set of results (Figure 5.5, Figure 5.6, Figure 5.7) show the contactless mechanocardiography signal synchronized with the ballistocardiography signal and the electrocardiography signal. The results also show that the second derivative of the signal taken from the contactless mechanocardiography device correlates to the BCG signal measured using the ballistocardiography device in the CIBER lab.

In the course of the experiment, all subjects were seated in a chair. The radar BCG signal was acquired by a microprocessor-based radar
data acquisition involved measurement of radar, ECG and respiration. All of the signals were acquired by a Biopac biological data acquisition system [13]. The block diagram used to derive the contactless mechanocardiography signal is shown in Figure 5.3

Figure 5.3. Three cycles of the signal acquired from the Doppler radar (bottom), the signal recorded directly from the sternum using an accelerometer (middle) and the lead I of the ECG signal
Figure 5.4. Experimental set-up of the Contactless Mechanocardiography Device in the CIBER Lab. The subject is sitting still, at a distance in front of the contactless MCG device. (Note that the setup shown in this picture is the prototype.)
Figure 5.5. Three cycles of the signal acquired from the Doppler radar (bottom), the signal recorded directly from the sternum using an accelerometer (middle), and the lead I of ECG signal during light breathing.
Figure 5.6. Three cycles of the signal acquired from the Doppler radar (bottom), the signal recorded directly from the sternum using an accelerometer (middle), and the lead I of ECG signal during maximum inspiration.
From these data, it is apparent that there is a close correlation between the signal acquired from the radar and the signal acquired from the accelerometer attached directly to the sternum, feeding the ballistocardiography device.

5.3. The contactless MCG results synchronized with ultrasound

The following are results derived from our experiment in which we synchronized an ultrasound and the contactless mechanocardiography device. Conducted at SFU’s Kinesiology Department, the subjects were lying down on a bed in a relaxed position. After gel was applied, the focused ultrasound beam was placed on the area that we wanted to
monitor. Simultaneously, the mechanocardiography device was pointed at the same monitoring location at a distance of 5 to 10 cm. The picture of the setup and the results are shown below.

Figure 5.8. Setup for monitoring arteries in left forehead area

Figure 5.9. Setup for monitoring basilar artery
Figure 5.10. Set up for monitoring left common carotid artery using the contactless MCG device and the ultrasound device at SFU's Kinseology Department
Figure 5.11. Carotid monitoring – First graph shows the signal taken from the ultrasound device and the second graph shows the signal taken from the contactless mechanocardiography device. ‘X’ axis stands for TIME and ‘Y’ axis stands for AMPLITUDE. As it is shown in the graphs, the signal taken from the ultrasound device (upper graph) does not show a meaningful signal but in the signal taken from the MCG device (lower graph) the pulses of blood flow movement can be seen as peaks in the signal.
Figure 5.12. Brachial monitoring - First graph shows the signal taken from the ultrasound device and the second graph shows the signal taken from the contactless mechanocardiography device. 'X' axis stands for TIME and 'Y' axis stands for AMPLITUDE. As it is shown in the graphs, in the signal taken from the MCG device (lower graph) the pulses of blood flow movement can be seen as high amplitude peaks in the signal.
Figure 5.13. Wrist monitoring - First graph shows the signal taken from the ultrasound device and the second graph shows the signal taken from the contactless mechanocardiography device. 'X' axis stands for TIME and 'Y' axis stands for AMPLITUDE. As it is shown in the graphs, the signal taken from the ultrasound device (upper graph) does not show a meaningful signal but in the signal taken from the MCG device(lower graph) the pulses of blood flow movement can be seen as peaks in the signal.
5.4. Results taken from the major arteries

The following results were taken by monitoring major blood vessels in healthy females. The signal from the left carotid artery and the right carotid artery were taken to make a comparison.

The veins and the ECG of the subject were measured simultaneously and a significant relationship was found between the two received signals. An ECG is commonly used to express the potential action of the heart. The signal received by the contactless MCG device from the veins shows the heart's activity, the level of blood lipids and the blood vessel condition.

The exact position in measuring cerebral cortical blood flow (CCBF) is between the sternocleidomastoid and the throat, at a level between the fourth and fifth cervical vertebrae.

During the experiment, all the subjects were sitting on a chair at a distance away from the contactless mechanocardiography device, and a physician in the CIBER Lab inspected all the results and measurements.
Figure 5.14. Oscilloscope screen-shot of common carotid artery, recorded from a female subject

Figure 5.15. Oscilloscope screen shot of carotid artery recorded from a female subject
Figure 5.16. Oscilloscope screen shot of foramen spinosum, recorded from a female subject

Figure 5.17. Oscilloscope screen shot of eye movement, recorded from a female subject
Table 5. The heart rate and respiration rate measurements using the radar MCG signal for eight subjects. The numbers represent the percentage of correctly detected heart beats or breathing cycles relative to the total number of beats or breathing cycles.

<table>
<thead>
<tr>
<th>Subjects</th>
<th>Heart rate</th>
<th>Respiration rate</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>93</td>
<td>90</td>
</tr>
<tr>
<td>2</td>
<td>-</td>
<td>94.2</td>
</tr>
<tr>
<td>3</td>
<td>84</td>
<td>90</td>
</tr>
<tr>
<td>4</td>
<td>91</td>
<td>90</td>
</tr>
<tr>
<td>5</td>
<td>100</td>
<td>87.1</td>
</tr>
<tr>
<td>6</td>
<td>89.4</td>
<td>92</td>
</tr>
<tr>
<td>7</td>
<td>100</td>
<td>100</td>
</tr>
<tr>
<td>8</td>
<td>-</td>
<td>87.5</td>
</tr>
<tr>
<td>Averages</td>
<td>92.9</td>
<td>91.35</td>
</tr>
</tbody>
</table>
Conclusion

We have introduced a methodology and design for a device that enables both monitoring and diagnosis of the heart and major blood vessels from a distance of up to 30 cm. We have also presented an overview of the device’s hardware and software, as well as the results of its clinical testing. The device does not need a connection to the subject’s body and is easy to use essentially anywhere. The device is designed to easily measure a subject’s mechanocardiography signal and therefore aid in the diagnosis of ventricular dysfunction as early as possible.

The clinical testing stage proved the accuracy of the MCG signal taken by the device. The MCG signal was extracted from the radar signal acquired by the radar-based MCG device. Then, by comparing it with two other heart-related-signals - BCG and phonocardiograph - we demonstrated that the extracted MCG signal corresponded to the mechanical functioning of the heart.

Two other medical applications of the new device, the detection of heart rate and respiration rate, were also tested using the radar MCG signal, in which an accuracy of 91.35% for respiration rate and 92.9% for heart rate were achieved. The accuracy of detection could be improved in the future by correctly targeting the MCG device on the sternum and through further hardware improvements.
Our clinical experiments have shown that, with such a device, we can not only monitor the mechanical movement of the heart without direct physical contact, but that we can also monitor the activity of major blood vessels.

As a potential future application, patients could wear the contactless mechanocardiography device in their clothing, without having to connect any sensors to the body. Another useful application of the device could be round-the-clock monitoring, which could be easily set up using a Bluetooth module to send the mechanocardiography signal to a PDA (personal digital assistant), cellphone or personal computer. The MCG signal data could be sent this way to doctors or other healthcare professionals, and could provide a comparative study of a few days' or weeks' worth of real-time data. With such a system, a doctor could see immediately when a patient is developing a problem. This could be very useful in the recovery of cardiac outpatients as well as in the diagnosis of sleep apnea.

Moreover, a heart patient that experiences chest pain could immediately send data coming from a contactless mechanocardiography device to his or her doctor. The doctor could then determine if the chest pain was a sign of heart attack or merely benign.

Essentially, fully utilizing the contactless MCG device in the field could enable vital health information to be relayed to medical professionals without the need for a hospital visit, either on a constant
basis, once or several times daily or simply whenever the relevant diagnostic data are required.
Appendix 1. Programming codes to control the LCD

00027 // Constants created by the qwizard and used for the UART communication
00028
00029 #define RXB8 1
00030 #define TXB8 0
00031 #define UPE 2
00032 #define OVR 3
00033 #define FE 4
00034 #define UDRE 5
00035 #define RXC 7
00036
00037 #define FRAMING_ERROR (1<<FE)
00038 #define PARITY_ERROR (1<<UPE)
00039 #define DATA_OVERRUN (1<<OVR)
00040 #define DATA_REGISTER_EMPTY (1<<UDRE)
00041 #define RX_COMPLETE (1<<RXC)
00042
00043 unsigned char peak = 100;
00044
00045 // USART1 Receiver buffer
00046 #define RX_BUFFER_SIZE 8
00047 char rx_buffer[RX_BUFFER_SIZE];
00048
00049 #if RX_BUFFER_SIZE<256
00050 unsigned char rx_wr_index,rx_rd_index,rx_counter;
00051 #else
00052 unsigned int rx_wr_index,rx_rd_index,rx_counter;
00053 #endif
00054
00055 // This flag is set on USART1 Receiver buffer overflow
00056 bit rx_buffer_overflow;
00057
00058 // USART1 Receiver interrupt service routine
00059 interrupt [USART1_RXC] void usart1_rx_isr(void)
00060 {
00061 char status, data;
00062 status=UCSR1A;
data=UDR1;
if ((status & (FRAMING_ERROR | PARITY_ERROR | DATA_OVERRUN))==0)
{
rx_buffer1[rx_wr_index1]=data;
if (++rx_wr Index1 == RX_BUFFER_SIZE1)
rx_wr_index1=0;
if (++rx_counter1 == RX_BUFFER_SIZE1)
{
rx_counter1=0;
rx_buffer_overflow1=1;
}
}

Get a character from the USART1 Receiver buffer

#pragma use`
char getchar1(void)
{
char data;
while (rx_counter1==0);
data=rx_buffer1[rx_rd_index1];
if (++rx_rd_index1 == RX_BUFFER_SIZE1) rx_rd_index1=0;
asm("cli")
--rx_counter1;
asm("sei") // allow interrupt
return data;
}

USART1 Transmitter buffer

#define TX_BUFFER_SIZE1 8
char tx_buffer1[TX_BUFFER_SIZE1];

#if TX_BUFFER_SIZE1<256
unsigned char tx_wr_index1,tx_rd_index1,tx_counter1;
#else
unsigned int tx_wr_index1,tx_rd_index1,tx_counter1;
#endif

USART1 Transmitter interrupt service routine

interrupt [USART1_TXC] void usart1_tx_isr(void)
00102 {
00103 if (tx_counter1)
00104 {
00105 --tx_counter1;
00106 UDR1=tx_buffer1[tx_rd_index1];
00107 if (++tx_rd_index1 == TX_BUFFER_SIZE1)
00108 tx_rd_index1=0;
00109 }
00110
00111 // Write a character to the USART1 Transmitter buffer
00112 #pragma used+
00113 void putchar1(char c)
00114 {
00115 while (tx_counter1 == TX_BUFFER_SIZE1);
00116 #asm("cli")
00117 if (tx_counter1 || ((UCSR1A &
00118 DATA_REGISTER_EMPTY)==0))
00119 {
00120 tx_buffer1[tx_wr_index1]=c;
00121 if (++tx_wr_index1 == TX_BUFFER_SIZE1)
00122 tx_wr_index1=0;
00123 ++tx_counter1;
00124 }
00125 else
00126 UDR1=c;
00127 #asm("sei")
00128 #pragma used-
00129
00130 // Standard Input/Output functions
00131 #include <stdio.h>
00132
00133 unsigned int read_adc(unsigned char adc_input);
00134 unsigned int adc_data;
00135 unsigned char newADC = 0;
00136 unsigned char prevADC = 0;
00137 unsigned int tmr0 = 0; // reset timer 0
00138 unsigned int time = 0;
00139 unsigned char buttons = 0xFF;
unsigned char freeze = 0;
unsigned char numSteps = 5; // display data every number of steps (increase to slow down)

// Button definitions identifying which PIIIN each button is connected to
#define BTN1 0x02
#define BTN2 0x40
#define BTN3 0x10
#define BTN4 0x80
#define BTN5 0x20

// Timer 0 overflow interrupt service routine
// In this function we check if a button is pressed and we reset the timer
// This timer is also used to keep track of a time period, which is used to clock and trigger the ADC

interrupt [T1M0_OVF] void timer0_ovf_isr(void)
{
    // Raise a flag which is used
    tmr0++;
    time++;

    // Check if button 1 is pressed - moves the peak line down
    if (((buttons & BTN1) == BTN1) && ((PIND & BTN1) == 0))
    {
        peak --;
    }

    // Check if button 2 is pressed - moves the peak line up
    else if (((buttons & BTN2) == BTN2) && ((PIND & BTN2) == 0))
    {
        peak ++;
    }

    // Check if button 3 is pressed - increases the rate of displaying
    else if (((buttons & BTN3) == BTN3) && ((PIND & BTN3) == 0))
    {
numSteps ++;
}

// Check if button 4 is pressed - decreases the rate of displaying
else if (((buttons & BTN4) == BTN4) && ((PIND & BTN4) == 0)) {
    if (numSteps > 0)
        numSteps --;
}

// Check if button 5 is pressed - freezes the monitor
else if (((buttons & BTN5) == BTN5) && ((PIND & BTN5) == 0)) {
    freeze = !freeze;
}

buttons = PIND;

// Reinitialize Timer 0 value
TCNT0=0x0F;

#include <delay.h>

#define ADC_VREF_TYPE 0x10

// ADC interrupt service routine triggered when the ADC is finished with the conversion and the data is ready to be read
interrupt [ADC_INT] void adc_isr(void)
{
    // Read the AD conversion result
    adc_data=ADCW;
}

// Read the AD conversion result
// with noise canceling
unsigned int read_adc(unsigned char adc_input)
{
    ADMUX=adc_input | (ADC_VREF_TYPE & 0xff);
00211  // Delay needed for the stabilization of the ADC input voltage
00212       delay_us(10);
00213
00214       #asm
00215       in  r30,mcucr
00216       cbr r30,__sm_mask
00217       sbr r30,__se_bit | __sm_adc_noise_red
00218       out mcucr,r30
00219       sleep
00220       cbr r30,__se_bit
00221       out mcucr,r30
00222       #endasm
00223
00224       return adc_data;
00225 }
00226
00227  // Function which formats and sends the data over the UART
00228  void outputData(unsigned int data)
00229 {
00230       char i = 0;
00231       char buffer[6];
00232
00233       // Formats and converts the ADC hex values to decimals
00234       // places the values in a buffer (array of characters) and terminates them with a '\0'
00235       sprintf(buffer, "%d," , data);
00236
00237       // Loop that goes through the buffer with decimal values and sends them one by one through the UART until '\0' is encountered
00238       while(buffer[i] != '\0')
00239          {
00240              // Send to UART
00241              putchar1(buffer[i]);
00242              i++;
00243          }
00244      }
00245
00246  // Function which initiates all the registers controlling the peripherals of the microcontroller
00247  void regInit()
00248 {
00249       // Input/Output Ports initialization
00250  // Port A initialization
00251  // Func7=In Func6=In Func5=In Func4=In Func3=In Func2=In
Func1=In Func0=In
00252  // State7=T State6=T State5=T State4=T State3=T State2=T
State1=T State0=T
00253  PORTA=0x00;
00254  DDRA=0x00;
00255
00256  // Port B initialization
00257  // Func7=In Func6=In Func5=In Func4=Out Func3=In Func2=In
Func1=In Func0=In
00258  // State7=T State6=T State5=T State4=0 State3=T State2=T
State1=T State0=T
00259  PORTB=0x00;
00260  DDRB=0x10;
00261
00262  // Port C initialization
00263  // Func7=In Func6=In Func5=In Func4=In Func3=In Func2=In
Func1=In Func0=In
00264  // State7=T State6=T State5=T State4=T State3=T State2=T
State1=T State0=T
00265  PORTC=0x00;
00266  DDRC=0x00;
00267
00268  // Port D initialization
00269  // Func7=In Func6=In Func5=In Func4=In Func3=In Func2=In
Func1=In Func0=In
00270  // State7=P State6=P State5=T State4=T State3=T State2=T
State1=T State0=T
00271  PORTD=0xFF;
00272  DDRD=0x00;
00273
00274  // Port E initialization
00275  // Func7=In Func6=In Func5=In Func4=In Func3=In Func2=In
Func1=In Func0=In
00276  // State7=T State6=T State5=T State4=T State3=T State2=T
State1=T State0=T
00277  PORTE=0x00;
00278  DDRE=0x00;
00279
00280  // Port F initialization
00281  // Func7=In Func6=In Func5=In Func4=In Func3=In Func2=In
Func1=In Func0=In
00282  // State7=T State6=T State5=T State4=T State3=T State2=T
State1=T State0=T
PORTF=0x00;
DDRF=0x00;

// Port G initialization
Func4=In Func3=In Func2=In Func1=In Func0=In
State4=T State3=T State2=T State1=T State0=T
PORTG=0x00;
DDRG=0x00;

// Timer/Counter 0 initialization
// Clock source: System Clock
// Clock value: 14.400 kHz
// Mode: Normal top=FFh
ASSR=0x00;
TCCR0=0x07;
TCNT0=0x0F;
OCR0=0x00;

// Timer/Counter 1 initialization
// Clock source: System Clock
// Clock value: Timer 1 Stopped
// Mode: Normal top=FFFFh
// OC1A output: Discon.
// OC1B output: Discon.
// OC1C output: Discon.
// Noise Canceler: Off
// Input Capture on Falling Edge
// Timer 1 Overflow Interrupt: Off
// Input Capture Interrupt: Off
// Compare A Match Interrupt: Off
// Compare B Match Interrupt: Off
// Compare C Match Interrupt: Off
TCCR1A=0x00;
TCCR1B=0x00;
TCNT1H=0x00;
TCNT1L=0x00;
ICR1H=0x00;
ICR1L=0x00;
OCR1AH=0x00;
OCR1AL=0x00;
OCR1BH = 0x00;
OCR1BL = 0x00;
OCR1CH = 0x00;
OCR1CL = 0x00;

Timer 2 initialization
Clock source: System Clock
Clock value: Timer 2 Stopped
Mode: Normal top=FFh
OC2 output: Disconnected
TCCR2 = 0x00;
TCNT2 = 0x00;
OCR2 = 0x00;

Timer 3 initialization
Clock source: System Clock
Clock value: Timer 3 Stopped
Mode: Normal top=FFFFh
Noise Canceler: Off
Input Capture on Falling Edge
OC3A output: Discon.
OC3B output: Discon.
OC3C output: Discon.
Timer 3 Overflow Interrupt: Off
Input Capture Interrupt: Off
Compare A Match Interrupt: Off
Compare B Match Interrupt: Off
Compare C Match Interrupt: Off
TCCR3A = 0x00;
TCCR3B = 0x00;
TCNT3H = 0x00;
TCNT3L = 0x00;
ICR3H = 0x00;
ICR3L = 0x00;
OCR3AH = 0x00;
OCR3AL = 0x00;
OCR3BH = 0x00;
OCR3BL = 0x00;
OCR3CH = 0x00;
OCR3CL = 0x00;
// External Interrupt(s) initialization
EICRA=0x00;
EICRB=0x00;
EIMSK=0x00;

// Timer(s)/Counter(s) Interrupt(s) initialization
TIMSK=0x01;

// USART1 initialization
Communication Parameters: 8 Data, 1 Stop, No Parity
USART1 Receiver: On
USART1 Transmitter: On
USART1 Mode: Asynchronous
USART1 Baud Rate: 115200
UCSR1A=0x00;
UCSR1B=0xD8;
UCSR1C=0x06;
UBRR1H=0x00;
UBRR1L=0x07;

// Analog Comparator initialization
Analog Comparator: Off
Analog Comparator Input Capture by Timer/Counter 1: Off
ACSR=0x80;

// ADC initialization
ADC Clock frequency: 115.200 kHz
ADC Voltage Reference: AVCC pin
ADMUX=ADC_VREF_TYPE & 0xff;
ADCSRA=0x8F;
}
// Constant keeping the value of the number of samples which are used to calculate the moving average
// This moving average smoothens out the signal representation
#define MOVING_WINDOW_SIZE 10

unsigned int mvTail = 0;
unsigned int mvHead = 0;

// Function for calculating
unsigned char mvCalc(unsigned char *bfr, unsigned char numItems)
{
    unsigned int i;
    unsigned int avg = 0;
    unsigned char mvAvg[MOVING_WINDOW_SIZE];

    // copy samples into temporary buffer
    for (i = 0; i < numItems; i++)
    {
        mvAvg[i] = bfr[i];
    }

    // add up all the values from the temporary buffer
    for (i = 0; i < numItems; i++)
    {
        avg += mvAvg[i];
    }

    // after all teh values are added up we divide by the total number of values to get the average
    avg /= numItems;

    return avg;
}

// detected pulses before bpm calculated
#define PULSES 3

void main(void)
{
    unsigned int data, i;
    unsigned char x = 0;
    unsigned char x2 = 0;
unsigned char mvAvg[MOVING_WINDOW_SIZE];
unsigned char avg;
unsigned int peakTimes[PULSES];
unsigned char peakHead = 0;
unsigned char peakTail = 0;
unsigned int sampleTime = 0;
char below = 1;
unsigned char step = 0;
char text[16];

regInit();

// Zero out the moving average temporary buffer
for (i = 0; i < MOVING_WINDOW_SIZE; i++)
    mvAvg[i] = 0;

// Initialize the LCD screen
LCD_Init();
// Clear the screen with white background
LCD_Clear(white);

asm("sei")
sprintf(text, "- SFU Hearth Beat -");
LCD_Print(text, 50, 80, 1, 1, 1, blue, white);
sprintf(text, "--- bpm"); // show on LCD

while(1)
{
    if (tmr0 > 0)
    {
        tmr0 = 0;
        putchar(newADC);
        // Read the ADC conversion results
data = read_adc(0);
sampleTime = time;
}
// Send the data over the UART
outputData(data);

// Shift the data 3 to the right in order to use the 7 most significant bits to fit the data in the display vertically
newADC = (data & 0x3FF) >> 3;

// record in buffer for moving average calculation
mvAvg[mvHead++] = data >> 2;

if (mvTail < MOVING_WINDOW_SIZE)
{
    mvTail++;
}
else
{
    mvTail = MOVING_WINDOW_SIZE;
}

if (mvHead >= MOVING_WINDOW_SIZE)
{
    mvHead = 0;
}

// Calculate the moving average
avg = mvCalc(mvAvg, mvTail) >> 1;

// The following if statements calculate the peaks by counting how many times the signal goes over and then under the horizontal cursor line, which is adjustable by the user
if (below && (avg > peak))
{
    below = 0;
}
else if (!below && (avg < peak))
{
    below = 1;
    peakTimes[peakHead++] = sampleTime;
}
else if (peakHead >= PULSES)
{
    peakHead = 0;
}

if (peakTail < PULSES)
{
    peakTail++;
}

if (peakTail >= PULSES)
{
    // collected enough peaks,
    calculate bpm
    unsigned int oldestTime,
    newestTime;
    unsigned int timeDiff;
    newestTime = sampleTime;
    oldestTime =
    peakTimes[peakHead];
    timeDiff = newestTime -
    oldestTime;

    // Assign the beats per
    minute value into the text variable so it can be displayed later on
    sprintf(text, "%03d bpm",
    (60000 / 16) / (timeDiff / (PULSES -1)));
    peakTail = PULSES;
}

// check if we got any pulses in
reasonable amount of time

// If the bpm is lower than 30 then reset
the bpm count
#define NO_PULSE_LIMIT ((1000/16) * 4)

if ((time - peakTimes[0]) >
    NO_PULSE_LIMIT)
{
    peakHead = 0;
    peakTail = 0;
time = 0;
peakTimes[0] = 0;
sprintf(text, "--- bpm");

if (step >= numSteps)
{
    // Display the data only if the freeze button is not pressed
    if (! freeze)
    {
        // Delete the old data
        LCD_Draw(x+1, 0, x+1, 132, white);
        LCD_Draw(x+2, 0, x+2, 7, blue);

        the left that represents the cursor pixel before displaying the new one
        LCD_Draw(x, 131-peak, 7, 131-peak, blue);

        data
        LCD_Plot(x, 131-(avg), green);

        prevADC, x+1, 131-newADC, red);

        prevADC = newADC;
        x++;
        if (x >= 175)
            x = 0;

        step = 0;
    }
    else
    {
        
    }

}
step++;  
  }  
  
  // If the freeze button is not pressed then display the beats per minute  
  if (! freeze)  
  {  
      LCD_Print(text, 108, 118, 1, 1, 1, blue, white);  
  }  
}  
}
Appendix 2. MATLAB script for the data communication

% Create a serial port object.
obj1 = instrfind('Type', 'serial', 'Port', 'COM4', 'Tag', '');

% Create the serial port object if it does not exist
% otherwise use the object that was found.
if isempty(obj1)
    obj1 = serial('COM4');
else
    fclose(obj1);
    obj1 = obj1(1);
end

% Set BaudRate at 115200.
obj1.BaudRate = 115200;

% Connect to instrument object, obj1.
fopen(obj1);

data=";
while isempty(strmatch('MCG',data))
data = fscanf(obj1);
end

% Parse the string to obtain coordinates
[lat,data] = strtok(data,',');
[lat,data] = strtok(data,',');
[nsCardinal,data] = strtok(data,',');
[long,data] = strtok(data,',');
[ewCardinal,data] = strtok(data,',');
lat = str2double(lat);
long = str2double(long);
% Display the results
disp({lat,nsCardinal,long,ewCardinal});

% Free the serial port
fclose(obj1);
Appendix 3. M-file used to show the contactless MCG signal on a real-time basis on the MATLAB program

function ScopeMath_Simple(useSimulatedData)

if ~exist('useSimulatedData','var')
    useSimulatedData = 1;
end

% GUI variables
hFigure = [];
hAxesRaw = [];
hAxesMath = [];
hStartButton = [];
acquiringData = false;
% Instrument control variables
interfaceObj = [];
deviceObj = [];
channelObj = [];
waveformObj = [];

% set up a timer to periodically get the data
% from the instrument
timerObj = timer('Period', 1.5, 'ExecutionMode', 'fixedSpacing', ...
                  'timerFcn', @getDataFromInstrument);

makeGUI();

if useSimulatedData
    msgbox('Using simulated data');
else
    connectToInstrument();
end
function connectToInstrument
    try
        interfaceObj = visa('tek', 'com4::1689::871::C010151::0::INSTR:');
        connect(deviceObj);
        channelObj = deviceObj.Channel(1); % read from channel 1
        waveformObj = deviceObj.Waveform(1); % default waveform measurement
    catch
        cleanupObjects();
        rethrow(lasterror);
    end
end

function getDataFromInstrument(hObject, eventdata)
    if useSimulatedData
        % 30 Hz sinusoid with additive Gaussian noise
        xData = linspace(0,1,256);
        yData = sin(30*2*pi*xData) + randn(size(xData))*0.2;
        xUnits = 'seconds'; yUnits = 'Volts';
    else
        if ~strcmp(deviceObj.Status, 'open') && strcmp(channelObj.State, 'on'))
            cleanupObjects();
            error('Not connected to device, or channel is disabled on the scope.');
        end
        [yData, xData, yUnits, xUnits] = ...  
        invoke(waveformObj, 'readwaveform', channelObj.name);
    end

% check the user closed the window while we were waiting
% for the instrument to return the waveform data
if ishandle(hFigure),
    axes(hAxesRaw);
    plot(xData,yData);
    xlabel(xUnits); ylabel(yUnits);
    %
    axes(hAxesMath);
[freq, fftdata] = powerSpectrum(xData, yData);
plot(freq, fftdata);
xlabel('Frequency (Hz)'); ylabel('Amplitude');
end
end

% the function takes the power spectrum of a waveform. X in seconds. Y is sample of waveform.

function [freq, fftdata] = powerSpectrum(x, y)
n = length(x); % number of samples
Fs = 1/(x(2)-x(1)); % sampling freq = 1/delta time
freq = ((0:n-1)/n)*Fs; % freq of power spec elements
fftdata = 20*log10(abs(fft(y))); % compute power spec
idx = 1:floor(length(freq)/2); % take half of the array
freq = freq(idx);
fftdata = fftdata(idx);
end

function makeGUI
hFigure = figure('deleteFcn', @figureCloseCallback);
 AxesRaw = axes('position', [0.13 0.51 0.775 0.31]);
 AxesMath = axes('position', [0.13 0.08 0.775 0.31]);
title('Raw Data');
 AxesMath = axes('position', [0.13 0.08 0.775 0.31]);
title('Processed Data');
 AxesRaw = axes('position', [0.13 0.51 0.775 0.31]);
title('Raw Data');
 AxesMath = axes('position', [0.13 0.08 0.775 0.31]);
title('Processed Data');
 hStartButton = uicontrol('Style', 'PushButton', ... 'String', 'Start Acquisition', ... 'units', 'normalized', ... 'callback', @startStopCallback, ... 'position', [0.70 0.84 0.18 0.06]);
set(hStartButton, 'callback', @startStopCallback);
end

function startStopCallback(hObject, eventdata)
if acquiringData
    if strcmp(timerObj.running, 'on')
        stop(timerObj);
    end
end
acquiringData = false;
set(hObject, 'string', 'Start Acquisition');
else
    acquiringData = true;
    set(hObject, 'string', 'Stop Acquisition');
    if strcmp(timerObj.running, 'off')
        start(timerObj);
    end
end
end
function figureCloseCallback(hObject, eventdata)
    cleanupObjects();
end
function cleanupObjects()
    if isValid(timerObj)
        stop(timerObj);
        delete(timerObj);
    end
    try
        if ~isNumeric(deviceObj) && isValid(deviceObj)
            disconnect(deviceObj);
            delete(deviceObj);
        end
        catch
            delete(deviceObj);
        end
        if ~isNumeric(deviceObj) && isValid(interfaceObj)
            fclose(interfaceObj);
            delete(interfaceObj);
        end
    end
    if ishandle(hFigure),

delete(hFigure);
end
end

end % of ScopeMath_Simple
REFERENCES


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