Feasibility of non-contact photoplethysmography

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FEASIBILITY OF NON-CONTACT PHOTOPLETHYSMOGRAPHY

BY

PECK YENG SHARON CHEANG

A Doctoral Thesis
Submitted in partial fulfilment of the requirements for the award of
Doctor of Philosophy of Loughborough University

June 2008

Supervisor: Professor Peter R. Smith and Dr. Sijung Hu
Department of Electronic and Electrical Engineering

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PECK YENG SHARON CHEANG, 2008
For Mom and Dad,

and in memory of Professor Peter Smith
ABSTRACT

This thesis explores and investigates the feasibility of a non-contact photoplethysmography system operating in both transmission and reflection modes. Several issues are addressed in the implementation of the non-contact system, including the dynamic range of PPG signals, ambient artefacts and effects of direct coupling, which is light that is detected without any interaction with the measured tissue area.

Plethysmography has been used in a range of biomedical applications to study blood volume changes. All current applications employ contact probes, where the transducers are positioned directly on the tissue surface. Non-contact measurements, where the transducers have no direct contact with the tissue surface, i.e. skin, are in demand for clinical benefits. Measurements by non-contact photoplethysmography can be obtained from any tissue since it is not probe limited, can be used on patients with burns as the probe does not touch the skin, and can reduce anxiety in patients, as they are not wired to any equipment.

A heuristic model for the non-contact transmission and reflection PPG systems was established to predict the effects of direct coupling and the designated experiments are performed to justify and interpret this model. The outcome from these experiments indicates that direct coupling can be affected by the light source position with respect to the position of the photodetector when using non-contact transmission PPG system. In order to compare the performance of contact and non-contact measurements, separate experiments were implemented to simultaneously compare the pulsatile and non-pulsatile signals obtained from these systems.

Non-contact reflection PPG system demonstrated that a quality signal can be recorded regardless of the tissue geometry in the fingers during measurement. Although a strong degree of correlation was obtained in each of the experiments for both the transmission and reflection mode PPG system, it is established that signal measured from non-contact reflection PPG has a stronger agreement with contact PPG measurements.
It is concluded that it is indeed feasible to measure PPG signals using both transmission and reflection modes of the non-contact system. With further development to enhance the current system, non-contact PPG could be a practical technology in future blood volume change related physiological measurement.
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<tr>
<td>ABPI</td>
<td>Ankle Brachial Pressure Index</td>
</tr>
<tr>
<td>APG</td>
<td>Air Plethysmography</td>
</tr>
<tr>
<td>CI</td>
<td>Confidence Interval</td>
</tr>
<tr>
<td>CPPG</td>
<td>Conventional Transmission Mode PPG</td>
</tr>
<tr>
<td>DISCO4</td>
<td>Discrete Sensing with Custom Optoelectronics</td>
</tr>
<tr>
<td>DPF</td>
<td>Differential Path Length</td>
</tr>
<tr>
<td>DVT</td>
<td>Deep Vein Thromboses</td>
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<tr>
<td>FDM</td>
<td>Frequency Division Multiplexing</td>
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<tr>
<td>Hb</td>
<td>Deoxyhaemoglobin</td>
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<tr>
<td>HbO₂</td>
<td>Oxyhaemoglobin</td>
</tr>
<tr>
<td>IPG</td>
<td>Impedance Plethysmography</td>
</tr>
<tr>
<td>₅ⁿ</td>
<td>Absorption Coefficient</td>
</tr>
<tr>
<td>₇ⁿ</td>
<td>Scattering Coefficient</td>
</tr>
<tr>
<td>LDI</td>
<td>Laser Doppler Imager</td>
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<tr>
<td>LED</td>
<td>Light Emitting Diode</td>
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<tr>
<td>NRPPG</td>
<td>Non-Contact Reflection Mode PPG</td>
</tr>
<tr>
<td>NTPPPG</td>
<td>Non-Contact Transmission Mode PPG</td>
</tr>
<tr>
<td>OSA</td>
<td>Obstrusive Sleep Apnoea</td>
</tr>
<tr>
<td>PAD</td>
<td>Peripheral Arterial Disease</td>
</tr>
<tr>
<td>Abbreviation</td>
<td>Description</td>
</tr>
<tr>
<td>--------------</td>
<td>-------------</td>
</tr>
<tr>
<td>PPG</td>
<td>Photoplethysmography</td>
</tr>
<tr>
<td>RlIV</td>
<td>Respiratory Induced Intensity Variations</td>
</tr>
<tr>
<td>SGP</td>
<td>Strain-Gauge Plethysmography</td>
</tr>
<tr>
<td>SNR</td>
<td>Signal to Noise Ratio</td>
</tr>
<tr>
<td>SpO₂</td>
<td>Arterial Oxygen Saturation of Blood</td>
</tr>
<tr>
<td>TDM</td>
<td>Time Division Multiplexing</td>
</tr>
<tr>
<td>TTI</td>
<td>Transthoracic Impedance Plethysmography</td>
</tr>
<tr>
<td>VCSEL</td>
<td>Vertical Cavity Surface Emitting Laser</td>
</tr>
<tr>
<td>VOP</td>
<td>Venous Occlusion Plethysmography</td>
</tr>
<tr>
<td>WPG</td>
<td>Water-Filled Plethysmography</td>
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CHAPTER ONE

1 INTRODUCTION
1.1 Thesis Overview

Plethysmography techniques have been used for decades as non-invasive diagnostic tools in vascular assessments. Restrictions in the use of contact probes is one of the major factors limiting the wide application of plethysmography. Photoplethysmography (PPG) is the only potential plethysmographic technique to employ a non-contact probe because it is based on the principle of light absorption properties of vascular tissue and does not directly depend on the physical changes in the limb. The aim of this thesis is to perform an initial feasibility study to record non-contact PPG measurements. The proposed technology would result in non-contact monitoring and assessments of the vascular system, thus widening the applicability of PPG technologies.

The first chapter introduces the circulatory system, focusing on the network of vessels within the human vascular system, their characteristics and their respective functions. Various plethysmography techniques used in clinical assessments are compared, analysed and the drawbacks will be highlighted. An in-depth review of the fundamental aspects of PPG including background physiology, light transport theory, generation of pulse wave and PPG waveform, modes of operation, breakdown of signal component, description of the arterial waveform and major applications of PPG are discussed. Limitations of current PPG probe are highlighted and the alternative use of non-contact probes in PPG is presented. The potential issues that might arise from non-contact measurement are discussed.

The second chapter contains an introduction to non-contact PPG and an in-depth review of the potential applications of the non-contact PPG system. A new sensor design and probe geometry arrangement for the proposed non-contact PPG probe operating in both transmission and reflection mode is described. The PPG platform used in the proposed non-contact system is also discussed. Issues relating to the non-contact PPG system, including the dynamic range, movement artefacts, ambient artefacts, and direct coupling effects are addressed.
The third chapter is dedicated to the development of the non-contact PPG heuristic model. This model aims to provide a simple description of the light that is transmitted through tissue and does not provide an in-depth analysis of the optical properties that is defined by the light diffusion and scattering that occurs in the tissue. A general heuristic non-contact PPG model, as a function of time and light source position, is described. A heuristic arterial model, which is influenced by the ratio of DC components originating from direct-coupled light and blood volume of non-pulsatile tissue, is presented.

Chapter four outlines the experiments that were conducted in the investigation of the feasibility of the proposed non-contact transmission and reflection mode PPG system. The protocols of the experiments and the data analyses are described in detail. Discussions on the experimental results, comparison with the prediction model and their implications is also included. The two main objectives of these experiments are to analyse the effects of direct coupling on PPG signals and to evaluate the feasibility of the proposed non-contact transmission and reflection PPG system through simultaneous comparison with the conventional contact PPG system. Two examples of PPG applications using non-contact PPG system are discussed.

The fifth chapter briefly reiterates the major achievements of the work and provides concluding remarks. Future work recommendations include further enhancements and extensions of the proposed non-contact system. Future clinical applications using the proposed system are also suggested in this chapter.

The references are collected in a common section following the main body of the report. As the thesis consists of coherent work on non-contact PPG, other work that were carried out in the research period are included in the appendices.

The author has been involved in the following publications:


Cheang P.Y.S. and Hu S., “Feasibility study of non-contact photoplethysmography”, to be submitted to Journal of Medical Engineering & Technology.

1.2 Vascular System

The human circulatory system comprises of three main parts, the blood, the vascular system and the heart. The main functions of the circulatory system are to:

- deliver oxygen to the cells in the body and remove carbon dioxide from them;
- carry nutrients to the cells;
- remove waste and poisons from the body that could be harmful if accumulated;
- help in the immune system of defence against infection;
- regulate body temperature by absorbing heat from the cells’ production of energy; and,
- transport hormones.

Blood is a liquid tissue that consists chiefly of plasma and three kinds of formed elements. Plasma is made up of water, but it also contains proteins, minerals and other substances. The three formed elements include red blood cells, white blood cells and platelets. Red blood cells carry oxygen into and carbon dioxide from the body, white blood cells help protect the body from disease, while platelets release substances that help blood to clot. Blood carries nutrients to the cells and waste products away. It also carries hormones and disease fighting substances to the cells in the body.
The heart, a hollow, muscular organ, acts as a pump to drive blood to the various organs in the body. The heart consists of two pumps that lay side-by-side. These pumps are in a relaxed state when blood flows into the heart and they contract to propel blood out to the body through the vascular system.

The vascular system in the human body is made up of a network of vessels, namely the arteries, the veins and the capillaries. The main function of these vessels is to transport blood through the body; spreading from the heart, oxygen and nutrients are delivered to every cell in the body, while carbon dioxide and waste materials are returned to the heart. The combination of the heart and the blood vessels are termed the cardiovascular system.
INTRODUCTION

Figure 1-1 shows the cardiovascular system that was first described by British physiologist William Harvey in 1628. The system forms a closed circuit whereby blood is pumped out of the heart through a set of vessels and is returned to the heart via a different set. This type of circulation is known as a closed-system circulation, as blood is kept circulating inside the vessels at all time. There are two circuits within this system, the pulmonary circulation and the systemic circulation.

In the pulmonary circulation, blood is pumped between the heart and the lungs only. Deoxygenated blood from the heart is pumped to the lungs. As blood flows through capillaries in the lungs, it picks up oxygen supply from the thin-walled air sacs that can be found at the end of the small bronchial tubes through the diffusion process. This freshly oxygenated blood then flows back to the heart. The systemic circulation then distributes oxygenated blood from the heart to all organs and tissues of the body, except the lungs, through the arteries. After the exchange process in the capillaries bed, the veins then carry the deoxygenated blood back to the heart.

Arteries (except the pulmonary arteries) carry the oxygenated blood from the heart while veins (except the pulmonary veins) carry back the deoxygenated blood. Capillaries form the interface between the arteries and the veins and it is here that the actual exchange of oxygen and nutrients occur. The pulmonary arteries transport the deoxygenated blood to the lungs while pulmonary veins transport back the oxygenated blood to the heart.

The main artery carrying oxygenated blood from the heart is called the aorta, which then branches further into large arteries, small arteries and arterioles. The arterioles link with the venules through the capillaries, which is the smallest vessel in the vascular system. The venules then form larger veins, and finally merge at the main vein vessel, known as vena cava, which carries blood back to the heart. This vascular system is shown in Figure 1-2[2].
Figure 1-2  Human Network of Vessels

The characteristics of these vessels change functionally and structurally with successive branching. Pressure in the vessels also changes along the circulation circuit, with the highest pressure detected in the arteries, followed by the capillaries and, finally, the veins. The reason for the decrease in pressure is due to the frictional losses (resistance) in the vessels as blood travels through the body.

Vascular disease includes any condition that affects the vascular system, ranging from diseases of the arteries and veins to blood disorders that affect the circulation system. Plethysmography is used in many applications as the non-invasive diagnostic tool for vascular diseases that are related to the arteries and the veins. The physiology of the vascular system, and some of the related diseases are discussed further in the following sub-topics.

1.2.1  Arteries and Arterioles

Arteries are vessels that deliver blood from the heart to the organs and the tissues in the body. The arterial system is a highly pressurised system that must withstand the pressure that is generated by the strong contractions of the heart. In order to accommodate and adapt to these pressures, arteries have thick walls that contain large quantities of connective tissues that are elastic and are surrounded by varying degrees
of smooth muscles that contract. The main artery of the heart is the aorta, which stems from the left ventricle of the heart, and then gives rise to various other large artery branches.

The presence of elastic connective tissues enables the arteries to stretch with each heart pulse and recoil back when the tension is released. Arteries have large radii, thus they serve as a low resistance conduit. As a result, arteries are specialised to serve as rapid-transit passageways for blood from the heart to the organs and tissues, and to act as a pressure reservoir to provide the driving force for blood when the heart is relaxing.

When an artery reaches an organ, it branches into numerous smaller vessels known as the arterioles. They have small enough radii to offer considerable resistance to blood flow and are, in fact, the major resistance vessels in the vascular system. The resistances of the arterioles supplied to each organ can be adjusted independently to accomplish two functions. Firstly, they are responsible for determining the relative distribution of blood flow amongst the organs, depending on the momentary needs of the body. Secondly, they help the regulation of the arterial blood pressure.

Unlike the arteries, the walls in the arterioles have very little elastic connective tissues and the amount of muscle decreases gradually from about three layers of smooth muscle cells to only one around the smallest pre-capillary arterioles. Tensions in their walls ensure that arterioles do not stretch under pressure. Instead, they act as a buffer between the arteries and the capillaries, shielding the delicate capillaries from the high-pressure arteries. At any point in time, the arteries and the arterioles contain 15 percent of the total blood volume in the body as shown in Figure 1-3.
Figure 1-3 Distribution Chart of Total Blood Volume in the Human Body

Figure 1-3 shows the distribution chart of the total blood volume in the human body. It is shown that most of the total blood volume is stored in the veins and venules, whilst the capillaries have the least amount of blood. The heart and the pulmonary vessels each has 8 percent and 12 percent of total blood volume at any one time.

1.2.2 Capillaries

Capillaries are the smallest blood vessels that link up the arteries and the veins. They branch extensively to bring blood within the reach of every cell in the body. Thus, the main function of the capillaries is the exchange of materials between the blood and the tissues. Capillaries provide nutrients and oxygen to the surrounding tissues, absorb waste and carbon dioxide, and excrete waste products from the body.

Capillary walls are very thin. They are composed only of a single layer of flattened endothelial cells, with no smooth muscles or connective tissues. This structure adapts them for the exchange, which is carried out through the process of diffusion. Due to the extensive branching of capillaries, the total surface area available for the exchange is very large and the distance between a capillary vessel and a cell is very close, thus resulting in a highly efficient exchange.
As the radii of the capillaries are very narrow, they offer considerable resistance to flow. However, as the surface area of the capillaries is huge, the total resistance is still much less than that of the arterioles. In spite of the large number of capillaries, they contain only 5 percent of the total blood volume at any one instance. Therefore, only a small volume of blood is exposed to an extensive surface area.

1.2.3 Veins and Venules

Blood leaving the capillaries enter the venous system, via the venules, into the veins to be transported back to the heart. The major function of the veins is to act as a low resistance conduit for blood flow from the tissues to the heart. Veins have large radii; therefore, they offer little resistance to blood flow. The total cross-sectional area gradually decreases as the smaller veins converge into progressively fewer but larger vein vessels. Consequently, the velocity of the blood flow increases as the blood vessels converge.

Compared to the arteries, the veins are less compliant, i.e. less capable to expand or contract, because they contain less elastic fibres and smooth muscles, thus supplying little elasticity. These result in the veins having little to no distensibility or stretchability and have little elastic recoil. These structural properties enable the veins to adapt to their second function of being a reservoir for blood. Under normal resting conditions, the veins contain approximately 60 percent of total blood volume.

1.3 Comparison of Plethysmography Techniques

Invasive measurements are usually the 'gold-standard' procedures for accurate diagnosis and evaluation of peripheral vascular disease. They are often expensive, demand time and expertise, as well as exposing the users to the possibilities of risks and complications. Plethysmography methods offer a non-invasive, risk-free
alternative, which can provide quantitative and qualitative physiologic information, and have been broadly studied and used in various biomedical applications.

Plethysmography is the method used to measure changes in the size and volume of organs and extremities due to fluctuations in blood volume. Plethysmograph is derived from the ancient Greek words of ‘plethysmos’ and ‘graph’. The word ‘plethysmos’ means “increase” with ‘plethys’ meaning “mass”, while ‘graph’ is the ancient Greek word for “write”. The main advantage of plethysmography methods is the ability to make non-invasive measurement, where no insertion or withdrawal of any instruments through the skin or body is needed during diagnosis. Hence, they are widely used as a screening platform to confirm clinical impressions before determining the need for further invasive testing.

Plethysmographic measurement can be recorded through either the direct or the indirect method. Changes in the blood volume as measured by the direct method are reflected directly by the transducers used, for example by a polyurethane cuff in air plethysmography (APG) and water-filled vessels in water-filled plethysmography (WPG). On the other hand, indirect methods evaluate the changes in certain properties of the body and then relate these changes to blood volume. The types of transducers used in indirect measurements include electrical impedance in impedance plethysmography (IPG), mercury-filled strain gauges in strain-gauge plethysmography (SGP), and optical transmission or reflection in photoelectric plethysmography (PPG). Each technique has its own advantages and disadvantages and Table 1-1 compares and analyses further the extent of differences between these plethysmographic techniques.

APG is one of the direct methods of plethysmography that provides an indication of the overall vascular function in the whole leg. Measurements are performed by encasing the leg, from the knee to the ankle, in a polyurethane cuff. Changes in the cuff represent changes in blood volume of the leg due to venous filling or emptying that is caused by the change in posture or exercise.
<table>
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<th>Properties</th>
<th>Method of Measurement</th>
<th>Commercial Availability</th>
<th>Specific Site for Measurement</th>
<th>Need for Skilled Technician</th>
<th>Ease of Use</th>
<th>Low Cost</th>
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<td>Impedance Plethysmography (IPG)</td>
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<td>Photoelectric Plethysmography (PPG)</td>
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WPG is another example of the direct method of plethysmography. It is used to estimate the total blood volume changes in the limb, usually in the forearm. The limb is placed in a vessel with water-tight seals, and the volume of water displaced from the plethysmography corresponds to the rate of change of blood volume of the limb.

Under a stable environment, changes in the blood volume measured by both the APG and WPG are directly correlated to the changes in blood filling, thus providing a simple and quantitative measure of blood volume in the entire limb or segment. Nevertheless, enclosing the limb in a ‘chamber’ for measurement increases the temperature difference between the room and the limb, which can result in a vasodilatory effect[^1], which introduces an error into the measurement system.

Locations for measurements using the direct methods are site specific due to the design of the cuff or vessel, thus limiting it to the limb, usually the forearm, thigh or calf. Compared to the other plethysmography techniques, APG and WPG are bulky and heavy and hence, they are difficult to transport. The enclosed cuff also causes immobilization to the patients for the duration of the measurement and can be difficult to mount. The applied pressure of the cuff can cause discomfort to the patient and thus, cannot be worn for long period. Both the APG and WPG required calibrations to be performed before every measurement and they need to be operated by skilled technicians.

Neither the APG nor the WPG is capable of performing arterial function analyses, which include arterial blood pressure and arterial compliance assessments, because the discrete arterial information cannot be extracted from the measured output. WPG is rarely available for commercial purposes, but it is still being used in laboratories for teaching purposes[^5] while APG is available commercially, albeit limited. Owing to these drawbacks, the APG and WPG are not the preferred methods.

On the other hand, the three indirect plethysmography methods, IPG, SGP and PPG, are often used in various vascular assessments. IPG is used mainly to assess deep vein thrombosis (DVT) using the venous occlusion technique. Electrodes are positioned onto the limbs to measure the impedance changes, where variations are an estimate of
the limb blood volume\cite{6}, and good correlation with the volume displacement plethysmography has been shown\cite{7}.

SGP can be used to provide an estimate of blood volume in the limb. It measures the percentage changes in the mercury gauge that encircles the limb. The percentage changes in the limb girth is used to deduce the percentage changes in the limb volume, based on the assumption that the percentage changes in the limb volume is twice that of the limb girth. A direct relationship between the resistance of the gauge and the circumference of the limb was established\cite{8} and this is comparable with the blood flow measured from WPG\cite{9}.

PPG is used to measure changes in blood volume, mainly in the peripheries. It measures changes in the light intensity that is reflected from, or transmitted through, the tissue from the optoelectronics that is positioned on the skin surface. The main application of PPG is pulse oximetry, which is used extensively to measure the arterial oxygen saturation of haemoglobin. Variations in the amplitude of the measured light intensity have a close correlation with the blood supply to the skin\cite{10}.

Indirect plethysmography methods are available commercially and are inexpensive compared to the existing ‘gold-standard’ imaging techniques. The equipment used in these indirect methods are compact and light, and hence, they are highly portable and practical. Although widely used, there are still some limitations to these indirect plethysmography methods.

SGP measurements are limited to the forearm and the lower leg due to the restriction of the mercury strain gauge. It cannot be used to measure blood volume at sites where the encircling gauge is not able to make full contact with the entire circumference of the body parts, such as in the ankle\cite{8}. On the other hand, the IPG and PPG methods are not confined as strictly to a particular measurement site and can be used on various body locations. However, it is more common that IPG and PPG measurements are performed in the lower limb and the periphery respectively.

Highly skilled technicians are essential to obtain accurate results when using IPG\cite{11} for vascular assessments. It is crucial that they master the positioning of the
impedance probe because different arrays of the probe can produce different outcomes. This procedure is more complicated compared to SGP and PPG, where probes positioning using these two methods are less critical and can be easily adjusted to obtain better readings.

Calibrations are necessary in all instruments to ensure accurate and precise measurements. SGP can be calibrated mechanically by adjusting a screw to deflect the gauge for a known distance$^{[8]}$, or electronically by simulating a length change using a known resistance change$^{[12]}$. PPG can be calibrated easily by adjusting the light intensity so that the detected light is above a predetermined baseline threshold. IPG is the hardest to calibrate due to the diversity in the placement of electrode arrays.

All the discussed plethysmographic methods utilise contact probes during measurements, where probes are positioned directly on the skin surface. The difficulties in probe placement, the limited sites of measurements and the dependency on skilled technicians or operators are identified as drawbacks related to using contact probes for measurements. Hence, there is a need to investigate other alternatives to overcome these drawbacks. An alternative solution is to use non-contact probes, where the transducer that is being utilised does not have any contact with the skin surface. Out of the analysed plethysmography techniques, the PPG method appears to be the only practical method that can use non-contact probes.

In both the APG and WPG methods, the cuffs need to encase the entire limb in order to acquire blood volume measurements. In IPG, electrodes need to be positioned directly onto the skin so that changes in the impedance can be measured. For accurate measurement in SGP, where changes in circumference are measured, it is necessary that the mercury gauge encircle the entire limb snugly. PPG, on the other hand, is based on the absorption properties of the vascular tissue when illuminated with light, and does not depend on the physical changes caused by the variations in blood volume. Hence, it is possible for the PPG method to adopt the use of non-contact probes. The main issue lies in the ability to provide adequate light illumination to measure the distance across the optoelectronics and the skin surface in order to produce accurate and assessable measurements.
1.4 Photoplethysmography (PPG)

Photoplethysmography is a non-invasive optical technique for measuring tissue perfusion. Firstly reported by Hertzman\textsuperscript{13}, it measures changes in blood volume indirectly based on the variations in light intensity passing through, or reflected from, skin tissue. It uses electro-optics as the source in acquiring the cardiovascular pulse wave that is found throughout the body. This pulse wave, known as the photoplethysmographic signal or blood volume pulse, is generated by the pulsation of arterial blood within the peripheral vasculature as stimulated by the quasi-periodic cardiac cycle.

An assumption implicit to the PPG method is that changes in the measured light intensity are due to changes in the blood volume. This technique is the preferred optical based technology in a number of biomedical monitoring applications as it is safe and easy to use.

1.4.1 Background Physiology

It is important to understand how human physiology may affect the PPG signal generation. The following sections describe both the cardiac cycle and the generation of a pulse pressure wave in the body.

1.4.1a Cardiac Cycle

The cardiac cycle, consisting of alternating periods of systole and diastole, is the period from one heartbeat to another heartbeat. Systole is the period of contraction and blood ejection from the heart, while diastole is the period of relaxation and blood filling. Identical events occur on both sides of the heart, except that the pressure is lower on the right side, as pressure is needed only to pump blood to the lungs. The heart structure is as shown in Figure 1-4.
During early diastole, the continuous inflow of blood from the venous system into the atrium causes the atrial pressure to be slightly higher than the ventricular pressure, even though both chambers are relaxed. This causes the blood to flow from the atrium into the ventricle, thus increasing the ventricular volume. Late in the ventricular diastole phase, the atrium contracts and fills the ventricle with more blood.

The ventricular diastole ends and the systole period begins at the onset of ventricular contraction. At this moment in the cycle, the ventricle will have its maximum amount of blood. When the ventricular pressure exceeds the aortic pressure, the ejection of blood begins and as the blood is forced into the aorta, the aortic pressure increases. The ventricular pressure decreases considerably as the blood is pumped out rapidly. However, usually only half the blood in the ventricle is pumped out. The remaining amount of blood is known as the end-systolic volume, which is the least amount of blood in the ventricle during the cardiac cycle.

At the end of ventricular systole, the ventricle starts to relax and the ventricular pressure falls below the aortic pressure, causing the aortic valve to close. This closure produces a discontinuity in the recorded PPG signal known as the dichrotic notch. No
more blood can leave the ventricle at this point as the valve has closed. When the ventricle pressure falls below the atrial pressure, the ventricular filling occurs once again, thus repeating another cardiac cycle.

1.4.1b Generation of Pulse Pressure Wave

During systole, blood is ejected into the ascending aorta through the semi-lunar check valves. This rapid rise in pressure at each systole expands the aorta, creating a pressure wave front that travels along the arterial system at about 700 cm/second\[^{15}\]. The pressure wave has an ascending and a descending limb as illustrated in Figure 1-5.

![Arterial Pressure Pulse Wave](image)

Figure 1-5 Arterial Pressure Pulse Wave\[^{18}\]

The pressure wave ascends steeply due to ventricular systole, i.e. when the vessel walls of the elastic arterial system expand to accommodate the extra volume of blood over the quiescent volume. This is followed by a gradual descent in the signal due to diastole, where the elastic arteries release the stored blood further into the vasculature as the heart valve closes, thus returning the blood volume back to its quiescent value.
The incisura or dichrotic notch found in the descending limb is caused by the sudden closure of the aortic valve during ventricle relaxation\(^{16}\) and by wave reflection\(^{17}\).

The peak arterial pressure is the systolic blood pressure, while the nadir is the diastolic pressure. The difference between the systolic pressure and the diastolic pressure is known as the pulse pressure. The mean pressure is the average pressure over the entire cardiac cycle and can be determined by integration of the pressure wave over time. In healthy young adults, the systolic pressure is 120\,mmHg and the diastolic pressure is 80\,mmHg. With advancing age, the systolic pressure increases to a greater extent than the diastolic pressure because of the loss of elasticity in the arterial vessels. Thus, a systolic pressure of 140\,mmHg and diastolic pressure of 85\,mmHg is not uncommon in those patients aged over 50 years old.

The arterial pressure wave changes its shape and amplitude as it travels further from the heart, which includes the loss of the sharp dichrotic peak, the increase in wave amplitude and the more obvious diastolic wave as seen in Figure 1-6. This alteration is caused by wave reflection, damping and the non-uniform elasticity of the vessels.

![Figure 1-6 Changes in pulse pressure along the arterial tree\(^{18}\)](image)

The rise in pressure during systole stretches the elastic walls of the arteries. As the vessel walls stretch, the kinetic energy is changed to potential energy and when the pressure falls, the elastic forces shrink the stretched wall. The potential energy changes back to kinetic energy and blood is driven in the direction where there is least
resistance to flow. The pressure wave does not fall to zero because the next phase of the cardiac cycle begins automatically at the end of the diastole phase as co-ordinated by a series of electrical impulses that are produced by specialised heart cells found in the sino-atrial node and the atrioventricular node.

Figure 1-7 shows that the pressure in arteries is on average about 120mmHg. Pressure from an external cuff generally needs to exceed this pressure before the arteries are fully occluded. Veins and venules have very small internal pressure, approximately 10mmHg and 25mmHg respectively. Therefore, an external pressure of 30 to 40mmHg is adequate to occlude fully the veins.

![Figure 1-7 Pressure of Vessels in the Systemic Circulation](image)

1.4.2 Transport of Light through Tissue

A photoplethysmography probe incorporates a light source and a photodetector, which are positioned on the skin surface, to detect the variations in light intensity that passes through tissue and blood of the vasculature. These variations arise from changes in the...
blood volume of the tissue and cause changes to the light attenuation. Light attenuation is a complicated process that includes absorption, multiple scattering, refraction and reflection.

When a beam of light reaches the skin, a small fraction of the light is reflected because of the difference in refraction index between the air and the skin. Light that penetrates into the tissue is attenuated by a variety of substances inside the tissue, and its propagation is governed by the absorption and scattering processes.

Absorption is due to specific compounds in the tissue, such as water and haemoglobin, and the absorption coefficient ($\mu_a$) is an indicator of the probability of absorption events per unit length (mm$^{-1}$) travelled. The absorption of light in a medium can be expressed using the Beer-Lambert law. For an incident intensity $I_0$, the transmitted intensity $I$ through the distance $r$ (mm) is affected by the absorption coefficient $\mu_a$ of the tissue.

$$ I = I_0 \exp(-\mu_a r) $$

The absorption length, which is the reciprocal of the absorption coefficient, is the distance required for the intensity to drop to $1/e$ of initial intensity value.

The absorption coefficient is linearly related to the concentration $c$ (mol) of the compound in the tissue, $\mu_a = \alpha c$, where $\alpha$ is known as the specific absorption coefficient. Each chromophore, i.e. a compound that absorbs light, has its individual absorption spectrum that represents the level of absorption at each wavelength. The level of absorption is thus affected by the concentration and by the wavelength dependent constant of the specific absorption coefficient of the chromophore. In a medium containing different chromophores, the overall absorption coefficient is the linear sum of each chromophore.

The above Beer-Lambert equation can also be expressed in base 10 logarithms instead of natural logarithms. When expressed in the base 10 logarithms, the absorption coefficient $\mu_a$ is known as the extinction coefficient $K$ and the specific absorption
INTRODUCTION

coefficient is known as the specific extinction coefficient. The specific extinction coefficient of oxygenated and deoxygenated haemoglobin\(^{19}\) is shown in Figure 1-8. Absorptions of both the haemoglobin types are high in the visible region but drop significantly towards the near infrared region, and at approximately 600nm, the absorption spectra between them remain significantly different. As the absorption in the near infrared region is weak, the illumination can traverse a greater distance in the tissue before it is completely attenuated. This enables deep tissue information to be extracted non-invasively\(^{20}\).

![Absorption spectra of oxygenated and deoxygenated haemoglobin](image)

*Figure 1-8 Absorption spectra of oxygenated and deoxygenated haemoglobin*

In clinical applications, the spectral region from about 600nm to 1300nm is of particular interest because the absorption of light in the biological tissues is at its minimum within this region. The upper wavelength limit for the spectroscopic techniques is dictated by the increasing dominant absorption of water, while melanin causes a high level of absorption at wavelengths shorter than 600nm. This absorption region demonstrates a “window” of transparency, also been referred to as the ‘therapeutic window’, as the wavelength of light within this region can penetrate to the order of centimetres into the tissue.
Scattering in the tissue originates from the inhomogeneities in the tissue as a result of the mismatches in the refractive indices that occur both between and within the cells. The density of the scattering particles and the cross-section of the scattering particles affects the degree of scattering, and the scattering coefficient ($\mu_s$) indicates the frequency between scattering events per unit length (mm$^{-1}$).

The characteristic scatter of the tissue is commonly expressed in terms of reduced scattering coefficient ($\mu'_s$), which corresponds to the effective number of isotropic scatters per unit length. The probability distribution function considers the probability of a photon being scattered in a given direction after each scattering event, and the phase function describes the probability of the scattering angle by using mean cosine angle to express the anisotropy of the scattered photon.

Anisotropy coefficient ($g$) ranges from -1 to 1, characterising the scattering events that are totally backscattering to those that are totally forward scattering. Isotropic scattering occurs when the anisotropy coefficient is zero, where the scattering events are uniformly distributed in all directions, i.e. becoming diffused. As the anisotropy coefficient in biological tissue is in the range of $0.69 \leq g \leq 0.99$\textsuperscript{[21]}, scattering events occur principally in the forward direction. This increases the optical path travelled by the photons and consequently, increases the likelihood for absorption events to occur.

The increased path length is longer than the physical separation of the optoelectronics, i.e. the actual distance separating the arrangement of the position of the light source and the photodetector. Thus, the mean optical path length (DP) is used as a measure of the average distance that light travels through tissue. It can be expressed as a product of the differential path length factor (DPF) and the physical distance between the light source and the photodetector. DPF reflects the adjustment in chromophore absorption due to the extended path length as caused by scattering, and is an indication of how much longer the light has travelled. It is a function of absorption coefficient, scattering coefficient, single-scattering phase function and geometry\textsuperscript{[22]}. It has been reported that the DPF decreases with decreasing scattering and increasing absorption\textsuperscript{[23]} and that the DPF increases gradually with increasing wavelength\textsuperscript{[24,25]}. 
The depth of light penetration into tissue is affected by the separation distance between the light source and the photodetector, and by the tissue optical properties. Large source-photodetector separation increases the influence from deep tissue on measured PPG signal\textsuperscript{[26]}. When the absorption is weak, light can penetrate deeper into the tissue, but as the absorption increases, the path of photon into tissue becomes flatter\textsuperscript{[27]}. The actual depth is very difficult to measure, and it varies with the concentration of haemoglobin and tissue thickness\textsuperscript{[28]}. The penetration depth is wavelength\textsuperscript{[33]} dependent as the absorption and the scattering in tissues vary with wavelength, and longer wavelength has deeper penetration into tissue.

The approximate depth of penetration is reported to be the distance in which the light intensity drops to a value of $1/e$ (37\%) of its initial intensity and a study has recorded the approximate penetration depth in fair Caucasian at various wavelengths\textsuperscript{[20]}. In a separate study, the diffusion theory is used to derive an empirical relationship for the mean penetration depth in terms of the separation distance of the optoelectronics ($r$), and the absorption and reduced scattering coefficients\textsuperscript{[29]}. This model predicts that the mean depth is dependent on $r^{1/2}$ and on $(\mu_a\mu_s')^{-1/4}$, and thus it increases with the distance between the optoelectronics and decreases with the absorption coefficient $\mu_a$ and reduced scattering $\mu_s'$.

Extracting values of the optical coefficients is a complex task and are affected by the conditions of the tissue. A comprehensive review of optical properties of various biological tissues have been compiled\textsuperscript{[21]}. Optical properties of blood are dependent on parameters such as haematocrit, osmolarity and haemolysis\textsuperscript{[30]}.

When the concentration of red blood cells is increased, the absorption coefficient reflects the change as $\mu_a$ is dependent on the concentration of the compound in the tissue and as there are more compounds in the blood, scattering events are also increased. Osmolarity is a measure of the hydration status in the tissue. In hypermolar state, the concentration of the haemoglobin in red blood cells will be higher as water diffuses out of the cells. The cells will shrink in size and this affects the refractive index and the internal reflections of the cells. The increase in scattering and internal
reflections results in higher $\mu_a$. Haemolysis is the breakdown of red blood cells when they die and the release of haemoglobin into the surrounding plasma. The scattering decreases as there are fewer red blood cells, and the haemoglobin in the plasma reduces the difference in refractive indices in the blood. Owing to the loss of membranes that provides internal reflection in the cell, the probability of absorption is reduced.

The volume of the tissue being probed is related to the depth of light penetration. As light penetrates deeper into skin tissue at longer wavelengths$^{[31]}$, more tissue volume will be investigated. At a shorter wavelength, light only penetrates the upper layer of the skin because light is strongly absorbed by bilirubin, oxyhaemoglobin and deoxyhaemoglobin$^{[33]}$. Figure 1-8 showed that absorption is low at wavelengths longer than 600nm, and hence, light is able to penetrate deeper into the tissue.

The detected light intensity also determines the volume of tissue being probed. As the illumination is absorbed by different chromophores, a higher light intensity increases the probability of light probing further into the tissue, and hence a larger pulsatile signal would be detected. In addition, the depth of penetration is the distance whereby the light intensity drops to 37% of initial intensity, and hence, the stronger the initial intensity, the deeper the penetration into tissue.

The light source-photodetector separation also affects the volume of tissue probed. The contribution from the deeper vasculature increases as the light source-photodetector separation increases$^{[26]}$. The light intensity however decreases with the increasing separation, as more light will be absorbed when it travels a relatively longer path length. Therefore, it is important to consider the limit of the light source intensity when deciding the optimum distance of the light-source separation.

The detection of direct-coupled light would also contributes to the amount of tissue volume probed. When this happens, the intensity of light that truly interacts with the tissue is reduced and this will affect light penetration into the tissue. Moreover, the PPG signal that is detected will not be reliable, as it does not represent the true physiological status of the tissue.
The sites of measurement also affect the volume of tissue being probed. When the skin tissue is thin and relatively transparent to light, e.g. finger and earlobe, light can easily penetrate into the tissue. When the skin tissue is thick, e.g. foot and forearm, more light will be absorbed by the thicker layer of the skin and hence, less light will reach the deeper layer of the tissue. Therefore, it is important to consider all the different factors before making measurements in order to maximise the volume of tissue probed.

1.4.3 Principle of PPG

In photoplethysmography, the illumination from a light source is emitted into the tissue, where it is scattered and absorbed. The light-tissue interaction is dependent on the wavelength of the light source. The influence of different wavelengths on PPG signals have been studied\cite{J2, J3} and it was shown that different wavelengths generated different PPG amplitude and different penetration into the vasculature.

1.4.3a Mode of Operation

Photoplethysmographic signal can be observed by illuminating a suitable pulsating vascular bed with an appropriate wavelength of light. As blood within the vascular bed pulsates with the cardiac cycle, the optical path length of the light source is altered accordingly, and consequently, a modulated signal is detected. Light in the near infrared region penetrates several millimetres\cite{J4} into tissues, where the dynamic absorption of the pulsating vascular bed modulates the detected light. Changes in this signal can be detected through both the transmission mode and the reflection mode probe.
1.4.3a1 Transmission Mode PPG

In transmission mode PPG, the light source and the photodetector are positioned on opposite sides of the vascular bed as illustrated in Figure 1-9. Blood and tissues within the vascular bed absorb light that is emitted from the light source, and the photodetector receives the remaining light intensity that passes through. This restricts the PPG measurement to anatomy where the tissue is relatively thin and has a high degree of superficial vasculature. The more common measuring sites of transmission mode PPG are the fingers, toes and earlobes in adults, and the foot and palm in neonates.

![Figure 1-9 PPG Probe in Transmission Mode](image)

1.4.3a2 Reflection Mode PPG

In reflection mode PPG, the light source and the photodetector are positioned adjacently in close proximity on the skin surface as shown in Figure 1-10. The illumination received by the photodetector is a result of the backscattered light that is returning from a range of depths within the highly scattering tissue.
An opaque shield is usually positioned between the optoelectronics to prevent any direct illumination from the light source to the photodetector without first passing through the tissue. This configuration enables measurement from multiple locations on the body that is not accessible using transmission mode PPG. For an accurate measurement, the probe needs to be attached firmly on the skin, and thus, the measurement sites needs to have a flat and wide surface area, e.g. on the chest, forehead and limbs.

1.4.3a3 Comparison between Transmission and Reflection PPG

The changes in the light intensity as measured by the transmission and reflection mode PPG are generally assumed to measure the changes in blood volume in the tissue. The PPG signal measured from both techniques are considered to be qualitatively the same. When the transmission and reflection PPG signals are measured simultaneously, an excellent agreement was found between the changes in the amplitude of the PPG signal\(^{[35]}\). Both signals can be quantitatively compared when the light-voltage relationship is the same between the two PPG modes\(^{[36]}\). The same algorithm is used for calculating oxygen saturation for both the transmission and reflection mode PPG measurements.
Optoelectronics of the transmission and reflection probes can be used interchangeably because the same type of light source and photodetector are used for both PPG modes. When the light source and the photodetector are positioned adjacent to one another, they act as reflection PPG, and when the distance between the optoelectronics is large enough that they are on opposite side of the tissue, they act as a transmission probe. The hardware and software used for signal acquisition and display can be the same for both transmission and reflection mode PPG.

The main difference between the two PPG modes is in the site of measurement. Measurement using transmission PPG is restricted to the peripheries where the tissue is highly perfused and relatively transparent to light illumination, whilst measurement using reflection mode PPG can be positioned on any flat surfaced anatomy.

Both the transmission and reflection PPG depends strongly on the optical properties of the tissue. In transmission PPG, the detected signal is a measure of light absorption by compounds in the tissue, such as oxyhaemoglobin and deoxyhaemoglobin. An arterial pulsation will produce an increase in light absorption. Light could easily penetrate through tissue of the peripheries because they are relatively transparent and have a rich supply of vessels.

In reflection mode, the detected PPG signal is a measure of backscattered light that emerges back from the skin and the intensity varies according to the anatomy at the site of measurement. Studies have also suggested that the orientation effect of erythrocytes may play a role in the origination of PPG signal as it could contribute to arterial opacity and would hence, affect the measurement of the PPG signal\(^{37,38}\). When the reflection probe is positioned on the forehead, the measured signal is greater because of the relatively thinner skin and the higher density of blood vessels. In contrast, the measured intensity is comparatively lower in limbs because of the comparatively lower density of blood vessels and the lack of close proximity to skeletal structure to help to reflect the incident light.

The reflection PPG signal is determined mainly by the superficial vasculature, whilst the transmission PPG signal indicates pulsation throughout the area between the
optoelectronics\textsuperscript{[35]. As a consequence, the PPG signal measured using the reflection mode is generally weaker and less intense compared to that measured using the transmission PPG\textsuperscript{[39,40]. Consequently, reflection PPG might be restricted to certain high perfusion areas.

The light source and the photodetector are generally housed in a reusable or a disposable probe. Reusable transmission probes are generally built as a clip where the spring of the clip extends appropriately to fit the tissue. The spring helps to maintain a uniform pressure on the periphery, with the applied force pre-determined by the probe manufacturer. The PPG signal recorded using the same probe would hence be free from variations in the applied probe pressure.

Disposable transmission probes and both reusable and disposable reflection probes require adhesive material to attach the optoelectronics onto the skin. Tautness of the adhesion determines the force that is being applied onto the skin. Measured signal can vary significantly depending on the pressure that is exerted by the probe onto the skin\textsuperscript{[41,42]}, as this can change the local blood flow in the tissue under the probe. When the pressure is too tight, the blood flow could be occluded, and when the pressure is inadequate, the probe might not have a proper contact with the skin.

Poor sensor-to-skin contact might cause the PPG signal to be more susceptible to motion and ambient artefact as the probe would not be fully covered by the tissue. Direct coupling could be detected when the photodetector is exposed to direct illumination from the light source. During its occurrence, direct coupling could cause a severe effect on the measured PPG signal\textsuperscript{[43,44], and when the exposure is too strong, PPG signal could be saturated.

1.4.3b PPG Waveform

The fundamental assumption implicit in PPG is that changes in the measured light intensity are caused by the corresponding changes in blood volume in the tissue under the probe. The use of a light source with a suitable wavelength ensures that blood volume changes are detected with a greater contrast than other physiological effects.
The significant relationship found between dynamic PPG signals and the dynamic strain-gauge plethysmography measurements\cite{45} provide strong evidence that the predominant cause of the dynamic PPG signal is a measure of the dynamic blood volume change.

Changes in blood volume and oxygen content during systole and diastole alter the path length of illumination in the tissue. In systole, the increase in oxygen content and blood volume increases the amount of oxyhaemoglobin pumped into the vasculature, as well as increases the optical path length in the tissue. These changes result in the increase of absorption of infrared light, and hence, a decrease in the total light intensity that reaches the photodetector. During diastole, the blood volume returns to quiescent values and as the absorption reduces, an increase in the light intensity is observed. Figure 1-11 demonstrates the changes in light intensity as measured using photoplethysmography.

![Intensity Plot of Photoplethysmography Signal](Image)

Other studies have demonstrated that the amplitude of the dynamic PPG signal is related to the variations in blood flow and volume throughout the cardiac cycle\cite{33,38}. However, no direct relationship has been formulated between the observed pulsations and the underlying physiological dynamics. The Beer-Lambert law is often used to aid
the understanding of optical transmission through tissue\textsuperscript{[46]}, with a broader theoretical applicability provided by diffusion theory\textsuperscript{[47,48]}.

Although the detected light intensity depends on many factors, both physiological and geometrical, the design of the optoelectronics with an appropriate control of light source intensity\textsuperscript{[49]} (compensating for skin absorption) can result in a quasi-static (DC) PPG signal that is dominated largely by the total illuminated tissue volume. The dynamic changes in the optical properties, such as the arterial pulsations, can be used to isolate the absorption by dynamic blood from the absorption by static tissue components.

1.4.3c Breakdown of Signal Component

The PPG signal can be broken down into two components; the rapidly alternating signal (AC component) which is caused by arterial pulsations, and the quasi-static signal (DC component) due to constituents, such as fluids, bone and tissue, which do not modulate light but have a fixed level of absorption. The pulsatile component contributes only a small percentage (1–5\%) of the total light intensity\textsuperscript{[50]}. This small change in the arterial blood volume will induce a parallel change in the optical path length, thus modulating the light absorption through the vascular bed as shown in Figure 1-12.

The arterial pulse amplitude has been interpreted physiologically as a measure of blood supply to the skin\textsuperscript{[10]}. Since the circulation of blood undergoes transition from high-pressure arteries to low-pressure veins, much of the non-pulsatile blood will be venous. Partitioning this non-pulsatile component due to various absorptions, i.e. the blood, skin, tissue and bone, is difficult. Nevertheless, this component on the whole has been found useful in comparative venous testing\textsuperscript{[51]}.

Electrical representation of the PPG signal consists of both AC and DC signals, where the AC signal is a dynamic component representing the arterial pulsations and the DC signal is a quasi-static component indicative of the venous blood volume.
1.4.3d Arterial PPG Waveform

The detailed analysis of the temporal and spectral characteristics of the arterial pulsation can result in some useful physiological information. For example, the contours of the arterial pulsation can be attributed to energy storage in the arterial vasculature\(^{159}\). The presence of a smaller peak on the volume waveform as shown in Figure 1-13, known as the dichrotic notch, is classically attributed to the closure of the aortic valve at the end of the ventricular systole\(^{160}\) and tends to diminish with age\(^{154}\). Low frequency trends of the PPG signal have been attributed to the effects of respiration and vasomotion\(^{155}\).

This dichrotic notch can move further from the diastolic pressure base into the systolic period of the next wave\(^{156}\), therefore needing some explanation other than the aortic valve closure. The nature of this is not clear, but can be explained by the concept of
wave reflection\textsuperscript{17}, where the increase in pulse wave velocity results in a faster reflected wave augmenting the forward wave.

![Diagram of arterial pulsations and dichrotic notch](image)

\textit{Figure 1-13 Shape of Arterial Pulsations in a PPG Volume Waveform}

Studies applying the use of various transfer functions have shown that there is a direct correlation between arterial pressure and arterial PPG\textsuperscript{57,58}. Qualitatively, proportionality between the two seems obvious, since any increase in pressure will cause vascular expansion, resulting in an increase in blood volume and thus, an increase in light attenuation. Therefore, it can be argued that tissues in the vascular bed directly affect the contours of the arterial PPG waveform, thus allowing the inference of vascular parameters by assessment of this waveform.

1.4.4 Major Applications of PPG

Owing to the non-invasive nature of PPG, various applications are being developed to ease and standardise the measurement of tissue perfusion. PPG is being used in many different environments within the biomedical field, e.g. in operating theatres, in clinics, and in ambulatory monitoring, for different purposes, including the monitoring and measurement of arterial and venous blood. These applications can use either transmission or reflection contact PPG probes to obtain the PPG measurements.
Pulse oximetry\(^{59}\) is currently the most dominant application of PPG. It measures arterial oxygen saturation in blood through relative absorption of haemoglobin and oxyhaemoglobin at two different light source wavelengths. PPG is also used in the study of peripheral vascular compliance\(^{60}\).

Arterial PPG can be used to examine physiological variables such as pulse and respiration rate\(^{61,62}\), blood flow\(^{38}\), arterial blood pressure\(^{63}\), and viscoelastic properties of the vessels\(^{64}\). Venous (quasi-static) PPG can be used in a variety of functional venous haemodynamic tests\(^{65}\). There have been renewed clinical interests in this technique due to advances in calibration\(^{69}\).

1.4.4a Pulse Oximetry

Pulse oximetry is the major application of PPG for measuring arterial blood perfusion. The oximeter was first used to measure oxygen saturation of blood (SpO\(_2\)), using different wavelengths of light on a heated earlobe with a continuously illuminated system operating in the transmission mode\(^{66}\). This method is based on two physical principles: fully oxygenated blood occurs only during arterial pulsations\(^{59}\), and oxyhaemoglobin (HbO\(_2\)) and deoxyhaemoglobin (Hb) absorb light to varying degrees depending on the light source wavelength\(^{67}\).

The chosen wavelengths must be in the spectra where the two haemoglobin species have different absorptions. Oxygenated blood has better absorption in the infrared region while deoxygenated blood absorbs better in the red light region, as seen in Figure 1-14. Thus, wavelengths of 660nm (red light region) and 940nm (infrared region) are commonly used in the commercially available pulse oximeters.

The illumination of tissue with two different light sources at different wavelengths produces a contrast that is dependent on the relative concentration of haemoglobin species, and therefore, the oxygen saturation. Consideration of source wavelengths on the opposite sides of the isosbestic point, which is the wavelength where the
absorption of both haemoglobin species are identical (800-815nm), will give the greatest contrast possible.

![Absorption Spectra of Deoxyhaemoglobin and Oxyhaemoglobin Showing the Two Most Commonly Used Wavelengths](image)

**Figure 1-14** Absorption Spectra of Deoxyhaemoglobin and Oxyhaemoglobin Showing the Two Most Commonly Used Wavelengths

The need for heated probes was removed when it was discovered that the absorbency ratio of PPG pulsations at the two wavelengths changed with arterial oxygen saturation\(^{46}\). This method used pulsations of the arterial blood to differentiate between blood flow and other tissue components that remain static, thus removing the optical properties of the skin, bone and tissue from the signal.

This then led to the development of pulse oximetry in 1980, which is an incorporation of PPG and oximetry in a finger probe operating in the transmission mode\(^{59}\). Since then, pulse oximetry has been used extensively in operating theatres worldwide to detect anaesthesia-induced hypoxia. There are many variants of pulse oximetry, such as reflection mode oximetry\(^{68}\), fibre-based oximetry\(^{69}\) and multi-wavelength oximetry\(^{70}\).
1.4.4b Venous Haemodynamic Testing

The main clinical applications of venous PPG are in the muscle-pump test and venous occlusion plethysmography. Both applications provide useful timing information on the timing of the venous refilling time, with additional information potentially provided by calibration of PPG signals\textsuperscript{[71]}.

1.4.4b1 Muscle Pump Test

The muscle pump test checks for venous blood displacement under active muscle conditions. Compression of the venous vasculature by the exercise process causes the displacement of venous blood, where the venous blood is literally squeezed from the venules into larger veins, causing the blood to flow towards the heart. This positive flow effort is termed as the muscle pump\textsuperscript{[72]} and it assists in the venous return to the heart, aided by the various venous valves that prevent a backflow of venous blood.

Venous insufficiency occurs when the venous return check valves are faulty, resulting in a venous backflow immediately after muscle pumping. This can cause venous oedema or pooling of the blood in the extremities, which is the excess accumulation of fluid in the tissue. Venous PPG monitors blood volume in the foot before and after the dorsiflex exercise. In healthy patients, this results in venous refilling by the arterial inflow alone\textsuperscript{[73]}, whereas patients with arterial insufficiency will suffer from shortened venous reflux\textsuperscript{[74]}.

1.4.4b2 Venous Occlusion Plethysmography

Venous occlusion plethysmography (VOP) is the test used to evaluate the functional properties of the venous system when the muscles are at rest. It is a measurement of changes in the tissue volume in response to a brief obstruction of venous return. The arterial inflow into the limb is unaffected, so volume in the limb segment increases in
correspondence to the rate of the arterial inflow. The changes in venous blood volume over time during VOP are illustrated in Figure 1-15.

![Blood Volume Changes during Venous Occlusion Plethysmography](image)

\[
\begin{align*}
\text{AF} &= \text{Arterial Inflow} \\
\text{VC} &= \text{Venous Capacitance} \\
\text{VO} &= \text{Venous Outflow}
\end{align*}
\]

**Figure 1-15  Blood Volume Changes during Venous Occlusion Plethysmography**

A pressure cuff is applied to the limb to occlude the venous outflow, resulting in an increase in blood volume due to venous pooling. At the start of the occlusion, the blood inflow is not hindered and the blood volume increases at a rate equal to the arterial inflow\(^{[75]}\), thus increasing the venous pressure. When this pressure rises above the transmural pressure, further venous filling is not possible and this leads to a plateau in the venous curve. The accumulated venous volume is known as the venous capacitance, which is a measure of the capability of the veins to accumulate blood volume at a set pressure. When the occlusion is removed suddenly, venous emptying may be studied. The velocity of venous outflow depends on the venous capacitance and the flow resistance of the venous system in the limb\(^{[76]}\).

VOP can provide quantitative information about blood circulation in the venous system, especially in the haemodynamic relevant thromboses. This test is used mainly to diagnose venous obstruction that may afflict patients with deep vein thromboses (DVT) in the thigh or pelvis. It is, however, more difficult to diagnose thromboses in
the lower leg as there are multiple veins that can compensate for any deficiencies in
the vein under test.

Under normal conditions, the pooled blood flows immediately out of the limb after
the occlusion cuff is deflated, hence the relatively high value of the venous outflow
parameter. When a significant delay is detected in the this parameter, the venous
return could be compromised and DVT is a possible diagnosis\cite{71}. Therefore, VOP can
be suitable as a quantitative diagnostic tool, as well as a monitoring tool for the
effectiveness of treatment.

1.5 Limitations of Current PPG Probe

There is an increasing use and research into utilising PPG in patient monitoring and
peripheral vascular assessments because of its non-invasive nature and ease of use.
Currently, all PPG applications use contact probes, operating either in the
transmission or reflection mode, to obtain measurements. These probes usually come
in both reusable or disposable form. There are however, several limitations to using
contact PPG probe, and these can be divided into three categories, namely the site of
measurement, contact force of the probe, and their use on wounds or damaged tissue.

A. Site of Measurement

Firstly, the use of the transmission PPG probe is limited to areas of the body such as
the peripheries and earlobes in adults, and the palm and foot in neonates. Owing to the
high vascular content, measurements are mostly performed in the fingers and toes. In
cases where a finger probe is utilised, measurements can only be obtained on the
peripheries that have a good anatomical fit to the probe.
B. Contact Force of the Probe

Another limitation that may affect PPG measurements is the contact force between the sensor and skin. Only a few studies have reported the effect of the force of the probe on PPG signal waveform and in each study the reflection mode probe was used. The pressure exerted on the skin by the sensor can influence the amplitude of the AC signal\[^{177}\]. Changes in the AC amplitude appear to have a similar changing trend, first an increase, and then a decrease with an increasing contact force\[^{41,78}\]. The initial increase could be caused by an improved sensor positioning on the skin, while a further increase in the pressure could restrict the venous blood in the tissue, and hence a reduction in the signal amplitude.

C. Wounds or Damaged Tissue

A contact probe cannot be employed to obtain measurements on skin with open wounds or in cases where there is external tissue damage, such as ulcers and burns, as this will cause an extreme distress to the patient. In addition, contact probes are difficult to sterilise such as in their use after detecting colonic ischaemia in abdominal aneurysm surgery\[^{79}\]. Moreover, the contact probe cannot be used for measurement when there is a need for mechanical isolation.

As a result, there is a need for research into an alternative method to overcome these drawbacks. A non-contact probe can provide such an alternative, as well as introducing new applications that were deemed unfeasible with the use of contact probes. Hence, the feasibility of non-contact probes is investigated further.

1.6 Non-Contact Techniques

A non-contact technique for mapping blood perfusion over a specific tissue area already exists. A Laser Doppler Imager (LDI) is intended for use in the study of spatial variations in tissue perfusion at discrete times. This technique is based on the
principle of laser Doppler shifts, where light hitting moving cells undergo a change in wavelength (Doppler shift) while remaining unchanged when hitting static objects. It is an established technique for an early assessment of burn depth\cite{80,81}. Research is being carried out to investigate the value of LDI in other clinical applications, such as in the imaging of plaque psoriasis\cite{82} and tumour tissue\cite{83}.

LDI scans the tissue sequentially, where the magnitude and frequency distribution are directly related to the concentration and velocity of the moving cells, but unrelated to the direction of movement\cite{84}. The laser light source is scanned over the tissue in about 5 minutes while a photodetector is used to detect the backscattered light from each point. When the scanning is completed, a colour-coded perfusion image is generated. Each colour corresponds to different levels of perfusion, defined as a fraction of the maximum perfusion level of the specific image. When comparing the perfusion between tissues, differences in colour and structure of the tissue optical properties should be taken into consideration.

The size of the area that is scanned is determined by the height between the light source and the tissue. The measured signal increases non-linearly when the height is changed over a range of 5cm to 20cm, as it reflects a reduction in the backscattered light signal. The LDI signal reduces non-linearly with increasing skin thickness, and thus the imager is more sensitive to superficial blood flow. The sensitivity of the flow measurement will decrease in an exponential manner as a function of vascular depth\cite{85}.

There are two commercially available LDIs, Moor LDI (Moor Instruments, Devon, UK) and Lisca PIM1.0 (Lisca Development AB, Linkoping, Sweden). The Moor imager uses a continuous motion scan, whilst the Lisca imager uses a step-wise scanning technique. Although the Moor imager offer a faster scanning rate, the relative movement between the light source and tissue could cause movement artefact that will reduce the quality of the image signal\cite{86}. On the other hand, the Lisca imager maintains the sensitivity of the image, but results in a longer scanning period\cite{87}. While maintaining a good sensitivity, rapid changes in blood flow over larger skin areas cannot be monitored. Although LDI is a non-invasive and non-contact
measurement technique, it does not provide good quality, real-time measurement of blood perfusion.

On the other hand, the proposed non-contact photoplethysmography, an alternative to LDI, can potentially provide an instantaneous, non-invasive and non-contact measurement, as initially investigated by Hayes[88]. Similar to other PPG devices, the photoplethysmographic signal can be displayed in real time, hence resulting in a quicker analysis and diagnosis. As the signal artefact can be visualised and identified immediately on the display monitor, steps can be taken immediately to rectify this problem in order to ensure a more accurate diagnosis.

1.6.1 Non-contact PPG

There are several reasons why the use of non-contact PPG is appealing.

1. Measurements can be performed from virtually any part of the body, as it is not limited to probe positioning, and hence increases the applicability of current PPG system.

2. The non-contact probe can be used when there is a need for mechanical isolation, for example, in operating theatres and in neonatal monitoring.

3. The use of a non-contact probe can reduce patients' anxiety in clinics and hospitals because they will not be physically attached to medical equipment or machine.

4. The use of non-contact probe will also remove any dependency on a skilled operator or technician.

5. Variations in PPG signal that are caused by the difference in tautness of adhesion and exertion pressure of contact probe on the tissue and by poor sensor-to-skin contact can also be resolved by using non-contact PPG probes.
6. As the non-contact probe will not be in contact with skin tissue, it reduces the potential risk of transmitting infections from patient-to-patient through the use of a contaminated probe.

The use of a non-contact PPG probe increases the applicability of current PPG applications because it is not constrained by the limitations faced by contact probes. Non-contact probes allow the technology to be applied in a broader context, and this is important in the pre-screening of patients to avoid unnecessary invasive assessments of vascular diseases, including their use in peripheral arterial disease and deep vein thrombosis.

In non-contact PPG, the light source and the photodetector are positioned some distance away from the vasculature. Owing to the potential increase in artefact interferences that might occur in the paths between the optoelectronics and vasculature, detection of the light intensity resulting from the tissue could be reduced. This reduction can cause a poor signal-to-noise ratio or, in worst-case scenario, no detection of PPG signals because of insufficient illumination.

This issue can be resolved using an array of light sources, instead of a single light source, to increase the amount of light intensity. This boost in illumination can compensate the effect of attenuation, as well as increases the number of interactions between light and tissue. In addition, reflectors can be used to guide and focus the light intensity, so that the light-tissue interactions, as well as the proportion of detected illumination due to tissue, can be maximised.

The design of a non-contact PPG probe design operating in reflection mode has already been developed and patented and this device can be used for monitoring physiological parameters such as oxygen saturation and pulse rate. Both the light source and the photodetector are not in contact with the tissue. Figure 1-16 shows the schematic representation of the non-contact reflection PPG as suggested in the patent.

Transmitted light that passes through the lens is focused on a selected region of the tissue. Scattered light that has penetrated across the tissue between the probes is then
re-emitted back through the lens to the photodetector. The lens at the photodetector acts as a barrier to block out shunted light, i.e. light that is detected directly from the light source without first interacting with the tissue, from interfering with the desired signal at the photodetector.

This patented device reports on the various methods of measuring reflection non-contact PPG using different housing and casing designs, where each design provides a built-in shunt-free, optical path through tissue. In this invention, the shunted light is avoided by keeping the light source and the photodetector separate by a light barrier, or by using mirrors to reflect the emitted light away from the photodetector.

1.6.2 Issues Arising from Non-Contact PPG

Several issues regarding the implementation of non-contact PPG have been identified, with some issues being more critical to the successful implementation of the device. These issues include the need to deal with the dynamic range of the PPG signal, ambient artefacts, movement artefacts and direct coupling.
A. Dynamic Range of PPG Signal

The PPG signal measured at the photodetector is dominated mainly by a quasi-static (DC) signal resulting from the absorption of static blood and tissue. Arterial pulsation, which reflects the dynamic changes in the optical properties of the tissue, are very small, typically 1-5% of the total light intensity\cite{50}. A suitable amplification technique is needed to cater for this issue, as it is crucial to provide a wide dynamic signal range.

B. Artefacts

Artefacts are any spurious signals or noises detected by the photodetector that could result in a poor signal-to-noise ratio, and could thus lead to misinterpretation of the PPG signal. Artefacts that occur during measurements are most destructive to the AC signal due to the relatively small pulsation size, although larger artefacts can also degrade the DC signal component. The two main types of artefacts are ambient artefact and movement artefact.

As the name suggest, ambient artefact is caused by the detection of an external light source that cannot be attributed to the effects of physiological origin, for example overhead lighting during surgery\cite{90,91}. A study showed that five of the light sources commonly used in the clinical settings do not significantly affect pulse oximeter readings\cite{92}. In current PPG applications, ambient artefact can be reduced by ensuring precise probe placement, additional shading at the measurement site, covering of the light sensor with an opaque shield, performing measurement under subdued lighting and by using electronic filtering. However, ambient artefact could pose a bigger problem in non-contact PPG settings because of the separation distance between the probe and the tissue.

Any physiological movements, voluntary or involuntary, that occur at the site of measurement can cause movement artefacts. Voluntary movement is caused by the conscious physical changes at the site of measurement, while involuntary movement is caused by the reflex mechanisms, such as twitching and shivering. The
consequence of movements is that they may cause severe artefact, which could alter the optical coupling in the tissue, as well as the redistribution of blood following limb movement; hence, corrupting the outputs of both the pulse rate and oxygen saturation in pulse oximetry\(^9\). These effects will be more significant in non-contact PPG system because the probe is more sensitive to tissue movements, especially to movements on the skin surface.

C. Direct Coupling

Direct coupling, or optical shunting, occurs when the photodetector detects light directly from the light source, with no interaction between the illumination and tissue. In conventional pulse oximetry using a transmission PPG probe, the effect of direct coupling is negligible because of the physical contact between probe and skin. This phenomenon will be more significant in non-contact PPG as there are possibilities that no light-tissue interactions will occur when light travels from the light source to the photodetector.

Direct coupling can occur during assessments of the extremities, where light can be coupled directly through the inter-digit gaps. Possible occurrence of direct coupling can also be observed in clinical applications where it is suggested that multiple PPG sensors are built directly into the fixtures. In these applications, the detection of direct coupling depends on the position of the non-contact probe and tissue when the measurement is taken. Although the optoelectronics are built-in with the measuring site on the body in mind, different body sizes and body positioning during the measurement may cause an exposure of the photodetector directly to the light source, thus detecting direct-coupled light that has not gone through the tissues of interest.

The critical issues identified in the use of non-contact PPG are movement artefact and direct coupling. Both of these issues may not occur in all situations, but when they do, they can cause significant impact on the PPG signal. In these situations, the measured PPG signal is corrupted and this may cause a high proportion of false alarms. In addition, the presence of these two issues will affect the ability of the PPG probe to obtain accurate measurements. The inaccuracies introduced by the incorporation of
such errors into the averaged value will then be more difficult to identify or clinically quantify. Consequently, there is a need to address these critical issues to address their impact on non-contact PPG measurements.

1.7 Thesis Aims and Objectives

Various plethysmographic techniques have been examined, and their drawbacks identified. The main aim of this thesis is to overcome the restrictions of conventional probes by investigating the feasibility of using non-contact probe in PPG systems operating in both transmission and reflection mode. With this in mind, the primary objectives of the thesis can now be stated.

1. Assemble a transmission and a reflection mode probe that enable non-contact measurements. The same probes will be used in all related experiments in this study unless stated otherwise.

2. Identify key issues that need to be addressed during the implementation of non-contact PPG system.

3. Develop a non-contact PPG heuristic model to describe the transmission of light through tissue. This model will be an expansion of the existing heuristic contact PPG model.

4. Develop a heuristic arterial model to observe the effects of direct coupling, based on the significance of light source positioning, on PPG signals.

5. Investigate the effects of direct coupling on PPG signals based on the significance in light source positioning and tissue geometry using non-contact transmission mode PPG. This will evaluate the prediction of the heuristic arterial model.

6. Investigate the effect of tissue geometry on PPG signals using non-contact reflection PPG.
7. Evaluate the feasibility of non-contact PPG measurements through comparison with the simultaneously acquired signals from a conventional PPG probe. This evaluation is conducted for both the arterial and venous components of the PPG signals.

1.8 Summary

The main function of the network of vessels in the cardiovascular system, namely the arteries, veins and capillaries, is to transport blood that carries oxygen and nutrients to the tissues, while taking carbon dioxide and waste products back to the heart. The basic functions of each of these vessels were presented. Vascular disease is any disorder that affects the circulatory system, and can be subdivided into arterial and venous disease.

Practically, photoplethysmography is the only plethysmographic technique that can adopt the use of a non-contact probe because this technique does not literally measure physical changes. Instead, it observes the attenuation in light intensity caused by variations in blood volume. Several background issues were reviewed to understand how human physiology and tissue optics, i.e. transport of light through tissue, can affect the PPG signal. Details on the principle of operation, as well as the major applications of PPG, were also discussed.

The non-invasive clinical monitoring technique of photoplethysmography has been presented. The transport of light through tissue has been reviewed. The propagation of light through tissue is affected by light attenuation and optical properties of tissue. Details on PPG signal generation, modes of operation, PPG signal descriptions as well as major applications have been discussed. Conventional PPG employs a contact probe, where both the light source and photodetector are positioned directly on skin surface.

Currently, all applications of plethysmographic methods employ contact probes as their sensor. The probe is positioned on or around the skin or limb, to investigate the
changes in blood volume. Several limitations in using contact probes were discussed and can be divided into three categories: the site of measurement, contact force of the probe, and measurements taken on wounds or damaged tissue. These problems, however, can be overcome by using non-contact probes.

Non-contact probes have the capability to overcome the drawbacks of conventional contact probes and various reasons for its appeal were highlighted. Non-contact PPG has the potential to improve and increase current plethysmographic applications.

As the principle operation behind contact and non-contact photoplethysmography is the same, the non-contact probe can be a substitute for a contact probe in current PPG applications. Several issues that need to be addressed when designing a suitable non-contact prototype, probe have been underlined. These issues include the dynamic range of the PPG signal, ambient artefacts, movement artefacts and effects of direct coupling. Concern regarding dynamic range and ambient artefacts can be resolved, but the remaining two issues are more critical. Movement artefact and direct coupling do not always occur, but when they do, the effects have the ability to cause low signal quality and even corrupt the PPG signal.
CHAPTER TWO

2 NON-CONTACT PHOTOPLETHYSMOGRAPHY
2.1 Introduction to Non-Contact PPG

The potential of non-contact PPG was initially explored by a Research Student predecessor in the Photonics and Health Technology Research Group at Loughborough University[88]. Preliminary work for non-contact PPG was suggested as one of the applications that would benefit from the non-linear artefact reduction system. In this early work, when the source power was set to a maximum with a 3cm distance between a reflection mode probe and a finger, the PPG signal could be obtained, but artefacts made it impossible to identify the valid arterial pulsations. This attempt encouraged further research into a practical non-contact PPG system that include the minimisation of direct coupling of light, maximisation of probe field of view, and maximisation of the amount of light detection.

The following sections will explore the feasibility of non-contact PPG, the design of the non-contact PPG probe, the use of the probe with a newly developed PPG system, and the various engineering issues involved. Possible applications of non-contact PPG are identified and advantages over the conventional contact probe are discussed.

2.2 Applications of Non-Contact PPG

Non-contact probes can be used as substitutes for contact probes in many of the current PPG applications. The contact and non-contact probes can be used interchangeably because the principle of operation between them is unaltered; the only difference is in the placement of the light sensor and photodetector when acquiring measurements.

Several challenges will need to be resolved before a non-contact probe can successfully replace a contact probe in clinical applications. The critical problem is movement artefact, as movement may corrupt the PPG signal, and hence the user will need to be stationary when the measurement is being recorded. In applications where the probe is to be in-built into the furniture at the clinics or another piece of equipment, the probe needs to be designed and positioned strategically so that it is
suitable for use on all subjects regardless of height and variations in body positioning. The wavelength and illumination power of the light source will need to be considered carefully. High power probes will provide better illumination, and hence provide better quality signal, but the amount of power must not be harmful to the tissue and must be within the accepted regulatory approvals.

2.2.1 Pulse Oximetry

The major application of PPG is pulse oximetry\textsuperscript{[59]} which is used especially in intensive care units and operating theatres to monitor arterial oxygen saturation of haemoglobin. A contact probe sensor is positioned on a relatively thin tissue bed, and the absorption ratio of red and infrared light passing through the tissue gives an approximate measure of the arterial oxygen saturation. In most clinical settings, oxygen saturation is monitored via a finger clip probe operating in transmission mode that is attached to the index finger. Non-contact probes can potentially replace conventional probes for monitoring oxygen saturation.

Contact probes can be obtrusive in the operating theatre, especially in surgery that is conducted on the upper body. Signal artefact can be generated when the connector that links the contact probe to the PPG system or the contact probe itself is accidentally moved during surgery. The additional force that might be exerted on the probe may cause physiological changes to the tissue, and hence affect the accuracy of the measurement. Non-contact PPG probes aim to reduce such occurrences and increase the reliability of the PPG measurement.

As the proposed non-contact probe need not be physically attached to the tissue, the probe can be positioned discreetly so that it will not be easily disturbed. Moreover, the optoelectronics can be encased in a solid casing that will protect the probe and resist movements due to external influence, thus avoiding any probe-induced artefact. The non-contact probe can also be used when there is a need for mechanical isolation during surgery, hence making it a valuable instrument to have in the operating theatres.
In order to prevent hospital acquired infections, it is important to use clean and uncontaminated equipment and medical devices, and this includes the use of PPG probes. When using reusable probes, a high level of disinfection practice needs to be adhered in order to yield a reliable decontamination of the probe. This is especially critical in the intensive care units when a patient’s defence system is compromised.

This adds to the operational cost of using the equipment, and there might be a risk that less than optimal decontamination procedures are utilised due to limited awareness or time restriction. This issue will be less critical when using non-contact probes since the probe does not have physical contact with the skin, hence reducing the potential risk of infections by contamination through medical equipment.

2.2.2 Measuring Physiological Parameters

Spontaneous fluctuations found in the heart rate, arterial blood pressure, and other parameters of the peripheral blood circulation system could be classified according to their frequency. These fluctuations are attributed to different activities of the two branches of the autonomic nervous system, i.e. the sympathetic and parasympathetic nervous system. The heart rate, pulse rate, and blood pressure in most patients increase under stressful situations, such as pain or anxiety, due to the activation of the sympathetic nervous system.

The use of non-contact probes aims to reduce such circumstances from occurring so that more reliable measurements can be obtained. Non-contact probes can be pre-installed into the furniture, for example, clinic seats, couches, medical tables and reception sofas, and measurement can be taken discreetly without the patients being aware of it. As a result, patients will not be wary of any perceived pain, or worry about feeling uncomfortable. This reduction in anxiety will hence increase the reliability and accuracy of the measurements that are taken.

Besides the heart rate synchronous variations, PPG signals also contain components that are synchronous with the respiratory variations, which are often referred to as
RIIV (respiratory induced intensity variations). The physiological background of RIIV is not fully understood, but it is suggested that the signal arises mainly from the variations in venous filling and arterial transmission. Studies of the RIIV have shown that low respiratory rates can be associated with more extensive pressure variations, that the respiratory volume can be related to the amplitude of the RIIV signal, and also that RIIV signals can be used to monitor respiratory rates. Respiratory rate is useful for the investigation in sleep studies, sports training, the evaluation of stress, as well as clinical applications.

2.2.3 Sleep Apnoea

Non-contact PPG can potentially be used to study obstructive sleep apnoea (OSA), which is a type of respiratory disorder that is caused by total or partial collapse of the upper airway. OSA is found to be associated with repetitive oxygen desaturation and fluctuations in certain haemodynamic parameters of the autonomic nervous system. Since PPG could be used to obtain both components of interest, the diagnosis of OSA could be performed at no additional cost.

Non-contact PPG probes could be fitted onto the user's bed, and as the probe will not be directly attached on the skin, the user will not feel any discomfort and there will be no risk of probe detachment during sleep. Multiple sensors would be positioned around the bed to cater for potential subject movement. The sensors would be illuminated at different positions on the body to enhance the probability of detecting a quality PPG signal. Hence, this enables continuous monitoring of the heart rate and respiratory rate to be acquired throughout the night.

2.2.4 Sports and Exercise

Non-contact PPG can be implemented in sports and exercise physiology for cardiovascular and respiratory monitoring, two of the important factors used to assess the efficiency of the performance and training. Non-contact probes can be adopted
to record various cardio-respiratory measurements without the hassle of mounting probes of different devices onto the athletes before and after training. In addition, non-contact probes can be installed on exercise machines in the gymnasium to monitor cardiovascular performance such as heart rate and cardiac output.

2.2.5 Vascular Diseases

Non-contact PPG can play a significant role in screening vascular diseases, which are any disorders that affect the circulatory system that is caused by the structural changes in the blood vessels outside the heart and brain. It is often the narrowing or clogging of the vessels that cause this change in structure, which ultimately causes a reduction in blood flow to the vascular tissues. This disorder is increasingly the focus of primary health care physicians and cardiovascular specialists.

2.2.5a Peripheral Arterial Disease

Peripheral arterial disease (PAD), also known as atherosclerosis, is categorised as the disorder caused by the peripheral arteries (blood vessels outside the heart) in the vascular system. PAD can affect any individual, but it is usually found in the elderly, and is more common and more severe in patients with diabetic problems.

The ankle brachial pressure index (ABPI) is an important tool for non-invasive diagnosis of PAD, assessment of the lower extremity functioning in PAD and risk stratification for cardiovascular events. ABPI measurement using the PPG technique is usually performed on the toe since it is less likely to be affected by the increased arterial wall stiffness as compared to the ankle[104]. When blood supply to the foot is severely reduced, the patient may develop ulcers, typically on the toe or heel or sometimes on the lower leg. The sore can be easily infected and quickly become serious, taking a long time to heal.
When an ulcer is developed at the extremity, it is not possible to attach the conventional PPG probe because of the pain that will be caused by the direct contact between the probe and the ulcer. In addition, inefficient disinfection of the reusable contact probe after each measurement could lead to the spread of infections between patients.

These problems can be solved with the use of the proposed non-contact PPG probe. Since the probe is not in direct contact with the skin, measurements can be performed easily without causing additional pain to the patients, and patient-to-patient infections can be avoided. Hence, the use of the non-contact PPG probe can be advantageous in large scale population screening where hygiene is a practical issue.

2.2.5b Venous Testing

The quasi-static (DC) component of the PPG signal is useful in venous haemodynamic testing. One of the main applications of venous PPG is the muscle pump test, which is a functional assessment of the leg vein system during exercise\cite{105}. A short venous refilling time, which is the time taken for the venous volume to return to normal after exercise, gives an indication of the incompetence of the venous valve in the lower extremity\cite{106}.

Non-contact PPG probes can be used to remove any hindrance to movement during exercise and to reduce the possibilities of inadvertent probe displacement or discomfort to the patient during the duration of the test. Measurements can also be taken immediately before and after exercise without having to readjust the position of the probe.

Another method of PPG-based venous haemodynamic assessment is venous occlusion plethysmography (VOP), which is performed by inflating an occlusion cuff at the extremity. The three main parameters of VOP assessments are arterial inflow, venous capacitance and venous emptying, which provide adequate approximations of the
overall changes taking place in the extremity\textsuperscript{[105]}. VOP is mainly used to diagnose venous obstruction that may occur in patients with deep vein thrombosis (DVT).

DVT is the clotting of blood that occurs mainly in the deep veins of the legs, and less commonly in the arm or pelvis. A short venous refilling time (less than 19s), measured after a dorsiflexion exercise, suggests the presence of deep venous insufficiency\textsuperscript{[107]}. In a study conducted using conventional contact mode PPG, limitations were found in groups of patients who were unable to conform to the exercise programme due to confusion, paralysis or in those with foot trauma\textsuperscript{[108]}.

Non-contact PPG can be used to conduct measurements on patients with paralysis or patients with foot trauma. The non-contact probe can be easily positioned on a more appropriate area of the limb, as the site of measurement is not limited by the probe design. Moreover, the non-contact probe will not hinder any movement of the limb during the exercise.

There is evidence that long-haul flights may increase the risk of developing DVT due to little or no in-flight exercise\textsuperscript{[109]}. The non-contact probe can be installed in the seats of aeroplanes to enable continuous monitoring of blood perfusion of in-flight passengers. An alarm indicator could alert cabin crew, or even alert the passengers themselves, when a reduced perfusion is detected so that early preventive steps can be taken. In-flight measurement can be recorded on the passenger’s bare foot or through the passenger’s sock using a non-contact probe operating at a longer wavelength of approximately 950nm.

### 2.2.6 Neonatal and Paediatric Monitoring

Besides being employed for clinical assessments and critical care monitoring in adults, PPG is also applied in the neonatal intensive care units to monitor the heart and respiratory rates in newborn infants\textsuperscript{[98]}. The observed signal gives important information and allows early prognosis of any cardio-respiratory insufficiencies.
Non-contact PPG can be a very useful tool in neonatal and paediatric monitoring, as attaching a contact probe to a neonate is often difficult due to their fragile nature, constant movement and also the probe attachment may cause discomfort. This problem can be solved by strategically positioning several non-contact probes directly onto the fixture of the bed or cot. Multiple positioning of the non-contact probes could ensure continuous monitoring and recording because when some probes detect movement artefact, other probes can still record a high quality PPG signal.

Hence, accurate recording of continuous measurements can be taken without having to attach any probe to the neonates. As the non-contact PPG system can be pre-built into the bed or cot, it can be used at home as an alarm system to reduce the rate of sudden infant death syndrome.

2.2.7 PPG Waveform Analysis

In many PPG applications where the focus is on measuring oxygen saturation and in-patient monitoring, little attention is paid to the pattern of the PPG signal. As the PPG waveform mimics an intra-arterial pressure wave, the shape of the arterial signal can provide valuable information regarding the cardiovascular system, the peripheral circulation, and the properties of the vascular network. Changes in the vascular properties as caused by vascular diseases or aging have been shown to influence the changes in the contour and timing of the PPG pulses\textsuperscript{[110,111,112]}. Similar to the conventional PPG probe, the non-contact probe can be used as a means to extract information from the PPG pulse wave. Signals acquired by the non-contact probe have the same features as those acquired by a contact probe: a systolic rising slope, a declining diastolic slope and in younger individuals, an obvious dichrotic notch. It should therefore be possible to extract feature information from PPG signals acquired using non-contact PPG probe.
2.2.8 Measurement on Wounded Skin

In addition to being used as a substitute to contact probe in current PPG applications, the non-contact probe can be useful in clinical applications that require no physical contact with the skin, for example on patients with skin trauma. Such patients have fragile skin and cannot tolerate the mechanical interaction required by the contact probe. This suggests that the proposed non-contact probe could be used for measurements on wounded tissue surfaces, such as on patients with burns and ulcers, as the probe is positioned away from the skin and has no physical contact with the traumatised tissue. In addition, measurements can be performed on patients with odd physiology such as missing digits, because it is not limited by the design of the probe.

2.3 Non-Contact PPG System

Limitations of the contact probe used in current PPG applications have been identified in the previous section. The non-contact probe may be used a replacement to improve the performance of, and to increase applications offered by, the conventional contact probe. As established, tissue perfusion can be measured using PPG probes operating in both transmission and reflection mode.

2.3.1 Sensor Design and Probe Geometry

In the standard pulse oximetry contact probe that is available commercially, the transmitter consists of one red and one infrared light source. Infrared and red wavelengths are used because light in these regions is absorbed at different extents by oxygenated and deoxygenated blood, hence creating a measurable contrast between the two.
In the proposed non-contact PPG system, the light source and the photodetector are positioned some distance away from the skin tissue, as illustrated in Figure 2-1. The path travelled by the non-contact PPG probe is longer compared to that of the contact PPG probe, as light must travel through the environment before interacting with the skin, and again through the environment before detection by the photodetector. Upon reaching the skin surface, the intensity might be too low to penetrate through the tissue or too low to generate a good signal-to-noise ratio signal.

![Figure 2-1 Arrangement of Non-Contact Photoplethysmography](image)

In addition, the gaps between the optoelectronics and the tissue surface can expose the light illumination to further interference. Hence, significant reductions in the intensity can occur whilst the light travels from the transmitter to the tissue, and from the tissue to the photodetector. In order to resolve these situations, a new probe design will be developed to enable more precise non-contact measurements.

2.3.1a Transmission Mode

For the design of a suitable non-contact probe, the chosen sensors must be easy to obtain, ideally monochromatic and low in cost. Standard light emitting diodes (LEDs)
are utilised in the construction of PPG probes because they fit the above requirements, and in addition, LEDs have low power consumption, are stable, are physically quite rugged, are small, possess a high radiance, and can be easily integrated electronically. LEDs emit incoherent narrow spectrum light when biased in the forward direction, and this is an effect known as electroluminescence.

It is essential to increase light illumination in non-contact PPG to cater for the increased path length and potential saturation by ambient light in order to supply sufficient intensity to generate good signal-to-noise signals. To achieve this, the non-contact transmission probe is designed using an array of three red and three infrared sources for illumination, as illustrated in Figure 2-2.

![Light Source Arrangement for Transmission Mode Non-Contact Probe](image)

The width of the designed transmitter is approximately 9mm and the sensors are arranged in such a manner that light from each wavelength can be evenly distributed onto the tissue. The arrays of the red and infrared light sources are surface mounted LEDs with a wavelength of 660nm (Model KP-1608SRT, Kingbright Elec. Co. Ltd.) and 880nm (Model KP-2012SF4C, Kingbright Elec. Co. Ltd.) respectively (see Appendix A and B). The surface mounted LEDs provide a large emitting area with a viewing angle of 120°, and hence allow light to be emitted in a wide field of view.
A red LED with a wavelength of 660nm was chosen because there is a comparatively large difference in the absorption between the oxyhaemoglobin and deoxyhaemoglobin, making it possible to detect small changes in the oxygenation of the blood. The infrared LED was chosen at 880nm due to the rather flat absorption spectra for both the oxyhaemoglobin and deoxyhaemoglobin in this region. The main advantage of this is that slight variation in peak wavelengths of the LED would not degrade the accuracy of the measurement and would make little difference to the calibration procedure.

The photon receiver consists of a surface mount photodetector (Model OSD15-5T, Centronic Ltd.) that has an active area of 15mm$^2$ and is able to detect wavelengths between 430nm and 900nm, thus accommodating the choice of the wavelengths used in the designed transmitter of the non-contact transmission probe (see Appendix C). The LEDs and photodetector chosen are easily available in the commercial market.

![Optical Reflector](image)

*Figure 2-3 Optical Reflector*

The non-contact probes need to be housed in a stable and secure position, and a set screw reflector is the option chosen, as seen in Figure 2-3 (Model reflector 1938, Carley Lamps, US) (see Appendix D). The reflector has a fine coating of aluminium oxide on the surface, which protects the reflector and maintains the brightness of the surface. The parabolic nature of the reflector makes it suitable for reflecting, collimating and refocusing visible, ultraviolet and infrared light. Other benefits of this
The parabolic optical reflector is suitable for the purpose of this research because the reflector collects light radiation at its focal point, and then reflects it as a collimated beam that is parallel to the axis. The design of the proposed non-contact probe using reflectors to focus illumination of light source and photodetector was aimed to maximise the probe's field of view.

The transmitter and the photodetector are each housed separately into a reflector, as they will be placed at opposite sides of the tissue bed, as seen in Figure 2-1a. The role of the reflector at the transmitter is to reduce spatial distribution of the LEDs, thus enhancing the amount of light that is focused on the tissue bed. In addition, the reflector provides some directionality of the illumination pattern onto the tissue.

![Figure 2-4 Parabolic Set Screw Reflector](image)

Figure 2-4 shows the reflector that is chosen for use in the design of the non-contact probe. This reflector has a diameter of 25mm and a ream of 9mm. The collimated area from the parabolic reflector can provide a uniform distribution over a wide range and can also maximise illumination across the tissue bed. The reflectors also provide a means to guide and focus illumination at both ends of the transmitter and photodetector. As the reflector has a ream of 9mm, in order to maximise the
reflections from the reflector, the optoelectronics were fabricated on a 9mm diameter circular board which is positioned at the base of the reflector.

Although larger reflectors are available, this reflector size was chosen because the illumination pattern of the light source when positioned 50mm away from skin surface has an approximate width of 50mm. This width is approximately the width of the tissue that will be investigated in this research. Thus, this increases the area of illumination and maximises the light-tissue interactions so that spatial averaging can reduce the observed physiological heterogeneity arising from subject movements.

As well as acting as a socket to hold the PIN photodiode, the reflector at the receiver is used to collect light from the tissue volume by directing, reflecting and focusing the illumination towards the photodetector, as illustrated in Figure 2-5. Moreover, by restricting the field-of-view of the PIN photodiode, detection of ambient artefact can be reduced, hence increasing the proportion of received illumination due to tissue interactions.

2.3.1b Reflection Mode

The probe designed for transmission mode is not suitable for use in the reflection mode probe. The intensity generated from the transmission probe was not adequate to
be used in this arrangement to produce good quality PPG signal, as signal measured from reflection mode PPG is generally weaker compared to that measured from transmission mode\cite{39,40}. Hence, it is essential to increase light illumination from the non-contact reflection probe.

The feasibility of the reflection non-contact PPG probe is highly dependent on the ability to detect a sufficiently strong reflection signal. A strong PPG signal is important in non-contact applications in order to sustain good signal-to-noise ratio, and to counter the potential increase in ambient and motion artefact.

Instead of using conventional LEDs, the vertical-cavity surface-emitting laser (VCSEL) which is a high precision optical sensor with low drive current, high-coupled power and narrow circular beam was opted for. In addition, this type of light source is easy to focus, stable and less sensitive to interference. Moreover, the use of lasers in medical devices have always been used with special precautions. However, VCSEL is not a pure laser light source, as it emits a mixture of coherent and incoherent light, and hence, it has some attributes of the LEDs. Therefore, VCSEL can be used with less precautions compared to laser light sources.

As in all commercially available reflection mode PPG, the light sources and the photodetector of the non-contact reflection PPG are housed together in the same probe. The designed probe is constructed using eight infrared VCSEL that operate at the wavelength of 850nm as illumination sources (Model VC850-TO46-LM, Higher Way Electronic Co. Ltd) (see Appendix E).

These light sources have good beam quality, low beam divergence and have a typical viewing angle of $10^\circ$. The narrow angle is suitable for this application as it is critical that the illumination from the light source is not detected directly by the photodetector. Although direct coupling in reflection PPG only happens occasionally, its occurrence could cause a severe effect on measured signal and hence, careful precautions have to be taken to avoid this.

The light sources are built uniformly around a high-speed silicon photodetector (Model S5821-03, Hamamatsu Photonics K.K., Japan) (see Appendix F) as shown in
Figure 2-6. The photodetector has a high-speed response, high sensitivity over a wide range of wavelengths, high performance and is available at low cost. It has an effective active area of 1.1mm² and is able to detect wavelengths in the range of 320nm to 1100nm, and hence, it comprises the range of light sources that is generally used in pulse oximetry probe.

The light sources and photodetector are mounted on a 23mm printed circuit board. Unlike the non-contact PPG transmission probe, the optoelectronics are not housed in a reflector. The main rationale behind this is that light illumination may be reflected directly from the reflector surface to the photodetector, resulting in the detection of direct-coupled light. Hence, to avoid this from happening, the printed circuit board is constructed on a flat surface area and is attached to a custom-made fixture.
2.3.2 PPG Platform

2.3.2a Methods of Ambient Artefact Reduction

Photoplethysmography platforms are easily attainable in the commercial market. These systems are ready to use, and can be easily connected to reusable or disposable PPG probes. Noise reduction system are usually incorporated into the PPG platform to reduce the effect of ambient light artefact.

Lock-in detection can be used to improve signal-to-noise ratio, as it is suitable for isolation of very small signals against larger noise background. It uses a technique known as phase sensitive detection to single out the component of the signal at a specific reference frequency and phase. Noise signals at frequencies other than the reference frequency are rejected and do not affect the measurement. For stable detection, the reference frequency must be the same as the input frequency. Both signals are phase-locked together so that they remain constant during measurement.

An effective detection of the small signal is determined by the cutoff frequency. A low cutoff frequency will improve the signal-to-noise ratio (SNR), but this requires longer settling time, thereby reducing the sampling rate. In contrast, a wider frequency band has a faster settling time, but at a reduced signal quality. The desired features of the lock-in detection are a short acquisition time and ability to measure low light levels as to provide large dynamic signal range, but these two are contradictory demands to one another.

Another technique to reduce the ambient artefact is frequency division multiplexing (FDM). Source intensity will be multiplexed at a frequency other than the expected frequency of ambient light. The signal that reaches the photodetector will be filtered and demultiplexed to extract the raw PPG signal. FDM however requires prior knowledge of the spectrum of the ambient artefact and changes in the frequency of the ambient light might not always be compensated in a fixed frequency FDM.

The most commonly used method is time-division multiplexing (TDM), whereby the LEDs are switched on and off at a fixed interval of time. The light reaching the
photodetector is associated with the measurement from the corresponding LED that is being switched on. When all the LEDs are switched off, the photodetector receives signals that are caused by ambient light. This ambient signal is subtracted from the signal received when the LEDs are on. TDM compensates for ambient artefact on a discrete basis, and is therefore not limited to any single frequency and is able to counter the effect of ambient artefact as it occurs. In addition, the subtraction process can be easily implemented electronically without neither compensating the sampling rate nor the SNR. As TDM does not compromise between sampling time and signal quality, it is thus the preferred method of ambient artefact reduction.

A band-pass filter allows signal between two specific frequencies to pass through while rejecting all frequencies outside these band. This filter is best placed after the processing of ambient artefact to separate the dynamic AC component from the quasi-static DC component of the PPG signal.

2.3.2b Discrete Sensing with Custom Optoelectronics (DISCO4)

The PPG platform employed is a commercially available self-contained technology known with the acronym DISCO4, which stands for Discrete Sensing with Custom Optoelectronics, by Dialog Devices Ltd, as shown in Figure 2-7. It is a technology with high sensitivity and is well suited for sensing small variations in measured signals when present with ambient light interference. This is vital since in non-contact PPG, ambient light interference is inevitable due to the separation between the probes and vasculature. The DISCO4 platform can support up to four light sources with different wavelengths.

DISCO4 can operate in both automated and manual intensity control mode. In automatic mode, the platform begins by sampling across the channels for a few cycles before adjusting individual intensity values to establish a suitable baseline. In manual mode, the user has complete control over the intensity settings. The level of intensity on each light source can be adjusted independently to suit the need of the user.
Photoplethysmography probes, both contact and non-contact, can be connected to the DISCO4 platform through the 9-pin D-type connector. This is inline with the standard used in commercially available pulse oximetry probes, and the pin-out is compatible with the Nellcor range of oximetry probes. The LEDs connected to the PPG probes are driven sequentially at $1\text{kHz/LED}$ with a constant current of up to $100\text{mA}$.

A) LED Driving Circuit

The DISCO4 platform uses programmable LED drivers to control the intensity of the light sources. Each light source is connected to a separate 8-bit LED driver so that the gain of each individual light source can be adjusted independently. The LEDs drive current is between 0 to $30\text{mA}$, therefore, the resolution is approximately $0.12\text{mA}$ per level.
B) Transimpedance Amplifier

The transimpedance amplifier amplifies and converts the low photodiode current into a corresponding voltage level. It has a high gain of 1MV/A, and hence it is suitable for amplifying the small AC signal that is detected when using the non-contact probe.

C) Ambient Subtraction

DISCO4 uses a built-in TDM unit to compensate for the ambient artefact that is detected during the measurement. The ambient light effect, sampled in a separate sensing period, is subtracted from the individual intensity samples to remove dependency from ambient light.

D) Band Pass Filter

The band pass filter has a bandwidth of 0.75 to 11Hz and is used to extract the very small dynamic AC component that is superimposed on the large DC component.

E) Variable Gain

The signal then goes through the programmable gain stages, between 15dB to 77dB, and has a gain resolution of 5 bits.

Figure 2-8 shows a sample of PPG signal recorded from the DISCO4 platform using a standard pulse oximetry finger probe with an intensity at level 32. The top trace portrayed the DC component whilst the lower trace displayed the AC component of the PPG signal. This sample demonstrates a strong PPG signal with good signal-to-noise ratio. The signal at the top of the graph represents the quasi-static DC signal (Volts). Small variations can be observed on this signal and they correspond to the changes in the pulsatile signal. The pulsatile signal represents changes in light intensity (Volts) due to changes in blood volume. The hardware specification for DISCO4 can be found in Appendix G.
2.4 Addressing Issues of Non-Contact PPG

In the proposed non-contact PPG probe, optical reflectors are used as a socket to hold the optoelectronics as well as to provide guides and focus for the illumination, while the DISCO4 platform is used to acquire the PPG signal. A data acquisition device, Labjack UI2 by LabJack Corporation, Colorado, USA, (see Appendix H) is used to interface DISCO4 with the computer and also to function as an analogue-digital converter. The digital output, obtained using LabView (Laboratory Virtual Instruments, National Instruments, Austin, TX, USA), is then analysed before presenting it to the user. However, several issues need to be addressed prior to the implementation of a non-contact PPG system.

2.4.1 Dynamic Range of PPG Signal

The common physical assumption implicit in PPG is that changes in measured light intensity are due to changes in blood volume. Appropriate sensor design coupled with
compensation for skin absorption by control of source intensity\(^{49}\) can result in a quasi-static (DC) PPG signal that is dominated mainly by blood and tissue absorption. Dynamic changes in the optical properties of the tissue, such as the arterial pulsations, can further be used to isolate absorption by dynamic blood from absorption by static tissue components. However, these changes are very small, so proper amplification is necessary to provide a wider dynamic range.

The commercial platform DISCO4 (Dialog Devices) has the flexibility to support applications that require extreme dynamic range and is especially suited for sensing small variations in signal in the presence of ambient light. The dynamic (AC) component of PPG signal is very small, encompassing only 1-5% of total light intensity\(^{50}\), therefore the DISCO4 platform offers a suitable means of recording.

**Figure 2-9 Controlling Dynamic Range of PPG Signals**

The dynamic range of the measured PPG signal can be adjusted as required using either the LED driver to control the intensity of the light sources, or the variable gain to control the magnitude of the signal output, or both, as shown in Figure 2-9. When the measured DC signal is low or below a preset threshold, the LED driver can be adjusted manually to achieve the desired illumination level. On the other hand, if the
dynamic AC component is weak, the gain control can be tuned to achieve the necessary amplification. Therefore, DISCO4 has the flexibility to support non-contact PPG applications that require extreme dynamic range.

2.4.2 Ambient Artefacts

In non-contact PPG, the effect of ambient light is more apparent than that in conventional PPG due to the separation of the probe from the tissue. This additional ambient detection can generate errors in the received signal. The inadvertent measure of ambient artefacts is not a theoretical problem in the recovery of PPG signals.

In the proposed non-contact PPG system, an independent measure of ambient effects is obtained using an additional sampling period in the time division multiplexing scheme, when all available light sources are extinguished. This effect may then be subtracted from the intensity samples during illumination to remove any associated dependency with ambient light.

2.4.3 Movement Artefact

Movement artefact is one of the critical issues encountered in the implementation of non-contact photoplethysmography. It can result in the detection of a poor signal or completely corrupt the output signal.

Many attempts have been made to analyse and reduce the influence of movements on the measured signal. Research has been carried out to quantify movement artefact in pulse oximetry, by using advanced filtering techniques to reduce the impact of the artefact. Movement artefact will continuously pose a problem in PPG applications, and thus far, studies have only managed to minimise, and not eliminate, the effect on the PPG signals\[^{113, 114}\].
In a conventional contact PPG probe, the predominant effect of movement artefact is caused by the optical coupling between the probe and the tissue\textsuperscript{[78,88]}, e.g. the degree of contact force exerted by the probe onto the tissue. On the other hand, movement artefact in the proposed non-contact PPG system is caused by the movement of the skin surface as well as the physical movement of the tissue under investigation. In order to counter the effect of movement artefact, all the subjects involved in this investigation were asked to remain motionless in a quiescent state whilst recordings were taken using the proposed non-contact PPG system. It was not possible to come up with an optimized solution as this study could not provide an in-depth understanding of the physiological effects of movement, and thus the nature of movement artefact remains not fully understood.

2.4.4 Direct Coupling

Direct coupling is the phenomenon where the photodetector detects light directly from the light source, without first interacting with tissue. There are two scenarios where direct coupling could occur: first, where light from the source is illuminated directly onto the photodetector; and second, where the photodetector is exposed directly to ambient light. Both of these usually occur as a result of inappropriate probe selection or probe misplacement\textsuperscript{[115]} and would significantly alter the accuracy of the measurement\textsuperscript{[43,44,116,117]}.

The potential of direct coupling occurring in conventional PPG probe is low, and can occasionally happen in reflection mode as the light source and the photodetector are positioned on the same side of the tissue. In commercial reflection probes, an opaque shield is usually built into the probe to counter this problem. In non-contact reflection probes, several sensor designs that can provide shunt-free optical path have been suggested\textsuperscript{[89]}.

In the detection of ambient light, pulse oximetry has recorded both falsely low\textsuperscript{[116]} and falsely high readings\textsuperscript{[118]} of oxygen saturation. When the probe was not properly
attached, falsely low or falsely high oxygen saturation values can be recorded as the measurement depends on the extent of detachment\textsuperscript{117}. In clinical practice, discrepancies in measurements could lead to incorrect diagnosis and treatment in patients.

Using a transmission contact probe, it is reasonable to assume that the effect of direct coupling is negligible, since human tissue physically separates the light source and photodetector, thus ensuring light-tissue interactions at all times. In addition, the optoelectronics are usually housed in a clip, hence ensuring that the probe would fit onto the tissue snugly.

Direct coupling is, however, not negligible in non-contact photoplethysmographic applications as there are always possibilities that light could be illuminated directly from the light source to the photodetector. Position of the body and limbs with respect to the non-contact probes would determine the amount of direct-coupled light that might be detected.

For instance, when measurements are performed on the finger, direct-coupled light will not be detected in conventional PPG because the probe is attached directly onto the periphery. Owing to the separation of optoelectronics from the tissue, illumination from the light source will be distributed over the tissue area of interest. The spread in light can increase the likelihood of light transmitted straight to the photodetector especially when the tissue area is smaller than the area of the light distribution.

Position of the body is significant in non-contact PPG applications that have sensors pre-built into the furnishings. Due to differences in body size, the tissue area of interest might not be positioned precisely between the sensors, hence enabling part of the illumination to travel directly to the detector without having prior interaction with the tissue. Although this situation does not always occur, the impact of this on the PPG signal could be disastrous.

In an idealised geometry configuration, polarising optics could be used to reduce the effect of direct coupling as the state of polarisation would be modified when it is reflected from or transmitted through a material. A polarising filter could be inserted
into the illumination path, between the light sensor and the tissue, while an analyser is positioned between the tissue and the photodetector. The analyser needs to be orientated so that only light that is orthogonal to the beam will pass through. Due to the properties of a polariser, light that penetrates through tissue will be depolarised as a result of diffuse scattering, while direct-coupled light would remain polarised. The output will clearly distinguish between the two different light attributes. Practically however, polarisation at different angles in the tissue caused by tissue inhomogeneity may alter the properties of light illumination. Multiple scattering of light at different angles in the tissue will be averaged out, and hence, polarisation is not an appropriate solution for eliminating direct coupling.

Direct coupling can be affected by the position of the light source with respect to the position of the photodetector and the tissue geometry where the measurement is taken. The effect of direct-coupled light on measured PPG signal can be represented in a heuristic model that will be discussed further in the next chapter.

It is predicted that the effect will be maximal when the light source and photodetector are aligned because the path between them is the shortest and therefore, the illumination will be strongest. This could cause an increase in the proportion of detected quasi-static PPG signal component, thus reducing the relative magnitude of the smaller pulsatile signal. When the optoelectronics are out of alignment, the light path increases and hence, will increase the possibility of light-tissue interactions. The increase in absorption will thus reduce the direct-coupled light that would be detected at the photodetector.

2.5 Summary

A variety of non-contact probe applications, ranging from pulse oximetry to skin trauma assessment and venous occlusion plethysmography, has been discussed and their benefits have been highlighted.
New non-contact PPG sensors, operating in both transmission and reflection mode, have been designed to cater for the increased illumination that is needed to overcome interference that might occur during the transmission of light. A commercial optoelectronics sensing platform, DISCO4 (Dialog Devices) was used because of its high sensitivity and suitability for sensing signals with small variations when ambient light interference is present.

Several issues relating to the implementation of non-contact PPG have been discussed. These issues are usually not critical in conventional PPG, but have higher importance in non-contact PPG measurements due to probe separation from the skin surface. The first issue discussed was the dynamic range of the signals, as PPG is composed of a very small AC component superimposed on relatively large quasi-static DC. Secondly, ambient artefacts can be damaging since large amounts of ambient light can be detected by the photodetector due to the separation between the skin and photodetector. Movement artefact also poses a problem, not only in non-contact system but also in conventional contact probe. Another issue covered is the effects of direct coupling, which only arise when the photodetector detects illumination directly from the light source.
CHAPTER THREE

3 PHOTOPLETHYSMOGRAPHY
   MODELLING
3.1 Introduction to Modelling

A model is a description of an analogy used to aid in the visualisation of an event that cannot be observed directly. It usually refers to mathematical equation(s) that can be applied to represent the system of interest. When a system becomes too complex, i.e. too difficult or too time consuming to model in its entirety, a simpler model is often used to predict part of the output, or to provide acceptable results within a reduced input parameter set.

Photoplethysmography measures blood volume change in the skin tissue, therefore, a model is used to illustrate and predict the complex light-tissue interactions. Two classical models exist for modelling interactions between light and matter, the Beer-Lambert law and diffusion theory. The Beer-Lambert law is a simplistic measure of optical density, assuming a classical, empirical interpretation of the illuminating light source. Diffusion theory on the other hand models the propagation of light in terms of diffusion of photon density as described by macroscopic absorption and scattering parameters.

The primary physical assumption of PPG studies is that changes in measured light intensity, transmitted through or reflected from the vasculature, are induced by corresponding changes in illuminated blood volume. The effects of blood volume changes may be modelled as modulations in the homogenous and macroscopic optical properties of the tissue. A heuristic PPG model was formulated based upon observations that certain aspects of physical behaviour can be described independently of the chosen model[119].

This heuristic model can be developed further to model the light-tissue interactions of non-contact PPG. The nature of the model enables quantitative descriptions of the characteristic behaviour in a general manner that is not constrained by the assumptions of any single physical model. In this initial stage, the heuristic model aims only to provide an intermediate step in proving the hypothesis that non-contact measurements are feasible by describing the transmission of light through tissue but it does not interpret the physical light-tissue interactions that occur.
3.1.1 Beer-Lambert Model on PPG

The Beer-Lambert law can be used to model the relationship of interactions between light source and tissue bed. It has been used to aid in the understanding of optical transmission through tissue\(^{46}\). The Beer-Lambert law states that the transmission of light through a turbid medium is a logarithmic function of the density or concentration of the molecules within the medium. In the context of photoplethysmography, this model can be used to describe the transmission of light through human tissue.

The intensity of transmitted light \(I\) is a function of path length on which the light travels, and the absorbance constant for a given particle at a given wavelength. Based upon Equation [1-1], the mathematical representation of the Beer-Lambert law is given by,

\[
\text{Beer-Lambert Law: } I(\lambda) = I_0(\lambda) \exp(-\mu_{\text{eff}}(\lambda) r) \tag{3-1}\]

where

- \(r\) is the path length of light travel (cm)
- \(\mu_{\text{eff}}(\lambda)\) is the effective absorbance as a function of wavelength (cm\(^{-1}\))
- \(I_0(\lambda)\) is the light source intensity
- \(I\) is the intensity of light transmitted (Wsr\(^{-1}\))

Equation [3-1] couples the path length and the effective absorbance as a function of wavelength into a single measure of optical density. The constant \(I_0(\lambda)\), has wavelength dependency to account for the discrete narrow bandwidth used by the light source. This equation is generally used as the basic theoretical understanding in pulse oximetry\(^{46}\).
The primary assumption of Beer-Lambert law when used in conjunction with PPG is that the optical path through tissue can be separated into static and dynamic components. The majority of light absorption is due to structural components of the vasculature and venous return, whilst the small dynamic absorption relies solely on the wavelength-dependent absorbance of the pulsating arterial blood. Hence, decomposition of the macroscopic optical density can be expressed by,

$$\mu_{\text{eff}}(\lambda) = \mu_b(\lambda)d(t) + \mu_t(\lambda)r$$

where

- $\mu_b$ is the effective absorbance of blood (cm$^{-1}$)
- $\mu_t$ is the effective absorbance of tissue (cm$^{-1}$)

Owing to variations in the pulsation of arterial blood, the path length associated with $\mu_b$ is denoted dynamically by $d(t)$ (cm) and is assumed identical for all wavelengths. As tissue is static and non-pulsatile, the light path it is represented by a static variable $r$ (cm).

Equation [3-3] is an expansion of the Beer-Lambert law using the decomposition expression found in Equation [3-2]. The overall received signal intensity can be shown as a product of two exponential terms, as seen in Equation [3-4].

$$I(\lambda) = I_0(\lambda)\exp\left[-(\mu_b(\lambda)d(t) + \mu_t(\lambda)r)\right]$$

$$I(\lambda) = I_0(\lambda)\exp(-\mu_b(\lambda)d(t))\exp(-\mu_t(\lambda)r)$$

The optical density of the received signal is wavelength-dependent, as shown in the previous equations. This dependency can be removed by assuming that the light source is of a single illuminating wavelength, thus
Equation [3-5], being a product term, is hard to implement. Various methods can be used to separate the multiplicative terms into a linear expression that would simplify implementation. The change in dynamic path length is very small, typically 1-5% of total light intensity.\(^\text{[50]}\). By using the truncated Taylor series, where \(\exp(x) \approx 1 + x\) when \(x\) is very small, Equation [3-5] can be re-written as

\[
I = I_o \exp(-\mu_o d(t)) \exp(-\mu_t r)
\]

The dynamic path length, \(d(t)\), can be either positive or negative, thus Equation [3-6] may be expressed as

\[
I \equiv I_o (1 - \mu_o d(t)) \exp(-\mu_t r)
\]

Equation [3-7] can be related back to the basic mathematical PPG signal representation, where the received light intensity, \(I\), could be expressed as a quasi-static component superimposed with the time varying pulsatile signal, i.e. \(I = DC + AC(t)\). The AC component in the detected signal can be extracted using filters. The ratio \(\frac{AC(t)}{DC}\) is expressed by

\[
\frac{AC(t)}{DC} = -\frac{I_o \mu_o d(t) \exp(-\mu_t r)}{I_o \exp(-\mu_t r)} \approx -\mu_o d(t)
\]

Equation [3-8], shows an expression that is directly proportional to the time-varying path length. This predicts that it is possible to obtain a quantity directly proportional to changes in path length caused by blood volume change.
3.2 The Heuristic PPG Model

The main assumption inherent in PPG studies is that measured light intensity, either transmitted through, or reflected from, a tissue bed, is induced by changes in blood volume. A heuristic model can be used to describe the effect of blood volume changes in the homogenous and macroscopic optical properties of tissue because of the fairly uniform distribution of blood in the vasculature. Certain aspects of the physical behaviour of PPG can be described independently from the chosen model, thus the behaviour may then be quantitatively described in a general manner\[119].

3.2.1 Definition of the Model

Measured intensity, resulting from light transmitted through or reflected from the anatomy, is dependent upon the wavelength and intensity of the light source, as well as any optical interactions with the subject. The primary assumption of this model is that received intensity can be separated into components originating from different optical paths, therefore identifying and specifying distinct optical effects.

When a number of light sources are used in conjunction with a single receiver to generate one or more PPG signals, the received intensity as a function of time, \( t \), and wavelength, \( \lambda \), is modelled by,

\[
I(t, \lambda) = \sum_{j=1}^{n} I_j(t) \left[ \alpha_j + \beta_j(t, \lambda) + \gamma_j(t, \lambda) \right]
\]

where

\( j \) labels the light source of intensity \( I_j \)

\( \alpha_j \) is the coefficient that represents the direct coupling between light source and photodetector
\( \beta_j \) is the coefficient that corresponds to light coupled via non-pulsatile tissue

\( \gamma_j \) is the coefficient that corresponds to light coupled via pulsatile tissue

\( \alpha_j, \beta_j \), and \( \gamma_j \) are coupling coefficients that depend on all geometric and spectral properties of source-detector positioning, artefacts, tissue dynamics and optical properties of blood, skin and tissue. The coefficient \( \beta_j \) includes coupling of light from any anatomical components that is not associated with dynamic arterial blood volume. Although the coefficient \( \beta_j \) does not correspond to the dynamic pulsation, this coefficient corresponds to quasi-static components, and hence it is a function of time. It is assumed that the effects of ambient light have been successfully eliminated by the time-division multiplexing technique.

The dynamic portion of the pulsatile tissue coupling coefficient, \( \gamma_j \), can then be attributed to an underlying pulsatile signal, \( p_j(t) \),

\[
\gamma_j(t, \lambda) = \gamma_j(\lambda) p_j(t) \tag{3-10}
\]

It can be assumed that the pulsatile signal represents a dynamic change in arterial blood volume and/or macroscopic optical properties of the tissue, which can be related back to a dynamic change in optical path length. This optical path length change causes the dynamic modulation of the coupling coefficient expressed in Equation [3-10] and therefore the observed intensity fluctuations in Equation [3-9].
3.2.2 Application

The developed model is underpinned by two fundamental assumptions:

1. The received intensity can be separated into components that originate from various optical paths, with the optical coupling efficiency being free to change between distinct paths.

2. The arterial pulsations may be modelled as a change in optical path length that ultimately modulates the coupling of light from pulsatile tissue.

Although the assumptions made are conceptually similar to those employed in the conventional interpretation of PPG signals, the generality is much improved by this heuristic model. This heuristic model may be used to describe both venous and arterial PPG signals, but it does not dictate physical interpretation of the signal, nor does it attempt to explain the complex light-tissue interactions.

The model given by Equation [3-9] is just a convenient method of describing the received intensity and its fluctuations. The coupling coefficients identify different optical path lengths with different optical coupling efficiencies, thus the received intensity can be separated into optical-path dependent components.

An analytic optical model that relates observation of intensity fluctuations to the origin of pulsations can then be applied to attribute the physical significance to these coefficients. Both Beer-Lambert and diffusion theory model have shown that the coefficients can be attributed to physical significance[119].

3.3 Heuristic Model for Non-Contact PPG

All models relating to photoplethysmography are a representation of light-tissue interaction and they each provide a convenient method of describing the observed intensity and its fluctuations. A heuristic PPG model[119] that can be used to express
the effect of blood volume changes in the homogenous and macroscopic optical properties was described in Equation [3-9]. This heuristic model can be extended further to include the proposed non-contact PPG system.

The primary assumption of the heuristic PPG model is that the measured intensity can be separated into components based on the difference in their optical paths, therefore distinguishing distinct optical effects. This assumption is applicable to non-contact PPG since the core of non-contact and conventional PPG is the same, which is to assess tissue perfusion by measuring changes in light intensity. However, in conventional transmission mode PPG direct coupling is negligible, but it is of significant importance in non-contact transmission PPG.

The model in Equation [3-9] can be expanded to describe non-contact PPG by taking into consideration the influence of light source position with respect to the photodetector. The measured light intensity, \( v(t,x) \) for non-contact PPG as a function of time, \( t \), and light source position, \( x \), for a single illuminating wavelength is given by

\[
[v(t,x)] = RI [\alpha(x) + \beta(x) + \gamma(x)p(t,x)]
\]

where

- \( t \) is time (s)
- \( x \) is the position of the light source (cm)
- \( R \) is the wavelength dependent responsitivity (A/W)
- \( I \) is the light source intensity (W/sr\(^{-1}\))
- \( p(t,x) \) is the time-varying optical path length (cm)
- \( \alpha \) is the coupling coefficient that represents the direct coupling between light source and photodetector (cm\(^{-1}\))
$\beta$ is the coupling coefficient that corresponds to light coupled via non-pulsatile tissue (cm$^{-1}$)

$\gamma$ is the coupling coefficient that corresponds to light coupled via pulsatile tissue (cm$^{-1}$)

$\alpha, \beta,$ and $\gamma$ are coupling coefficients that depend on all geometric and spectral properties of the light source and photodetector positioning, artefacts, tissue dynamics, and both skin and tissue optical properties.

The coefficient $\alpha$ is a constant (DC) component, which represents light that does not interact with any tissue component, but is detected directly from the light source. The non-pulsatile coefficient, $\beta$, (DC component of blood and tissue) depends on tissue optical characteristics and includes all effects not attributable to arterial pulsations. Thus, the pulsatile coefficient, $\gamma$, (AC blood modulation) represents only the coupling of arterial pulsations, which would normally be much smaller than the corresponding $\beta$ value.

All three coefficients identify different optical paths, thus separating the detected light into different optical path dependent components. As the measured light intensity is composed of a small dynamic signal superimposed on a large quasi-static DC, the heuristic model in Equation [3-11] can be used to describe individually each of these components.

As this thesis focuses on the development of the heuristic arterial model of the non-contact PPG system, it did not explore the development of the heuristic venous model. Separate research in the Photonics and Health Technology Research Group at Loughborough University has been performed to carry out a more in-depth investigation into venous oximetry and the development of venous modelling.$^{[120]}$
3.3.1 Heuristic Arterial Model

Rapid modulations in the measured light intensity are caused by arterial pulsations and are represented by the dynamic component in the model. This dynamic portion in the signal can be related back to the dynamic change in optical path length, which is a quantity that is proportional to the ratio of pulsatile to non-pulsatile tissue components. Therefore, the dynamic signal can be expressed by

$$[3-12] \quad ac = \frac{AC(t,x)}{DC(x)} = \frac{\gamma(x)}{\alpha(x) + \beta(x)} p(t,x)$$

It is observed that this equation yields a dynamic term that is independent of light source intensity. Normalising the arterial pulsation with respect to the position, $x_0$, where the light source and photodetector are aligned gives,

$$[3-13] \quad \hat{v}_{AC}(t,x) = \frac{\gamma(x)p(t,x)}{\alpha(x) + \beta(x)} \frac{\alpha(x_0) + \beta(x_0)}{\gamma(x_0)p(t,x_0)}$$

where $\hat{v}_{AC}$ is the normalised AC signal.

Factoring and simplifying Equation [3-13] gives,

$$\hat{v}_{AC}(t,x) = \frac{\gamma(x)}{\beta(x)} \frac{\beta(x_0)}{1 + \frac{\alpha(x_0)}{\beta(x_0)}} \frac{\beta(x_0)\gamma(x_0)}{\beta(x)} \frac{\gamma(x_0)p(t,x_0)}{\gamma(x)p(t,x)}$$

$$\hat{v}_{AC}(t,x) = \frac{1 + \frac{\alpha(x_0)}{\beta(x_0)}}{1 + \frac{\alpha(x)}{\beta(x)}} \frac{\beta(x_0)\gamma(x_0)}{\gamma(x)} \frac{p(t,x_0)}{p(t,x)}$$
where $\hat{v}_{AC}$, $\hat{\beta}$, $\hat{p}$ and $\hat{\gamma}$ represent the normalised value.

Measured light intensity from direct-coupled light is lower than non-pulsatile tissue components when the majority of the light is illuminated on the tissue. Therefore, when $\alpha << \beta$, a first order representation of Equation [3-14] is

$$\hat{v}_{AC}(t,x) = \frac{\gamma \hat{p}}{\beta} \left( \frac{1 + \alpha(x_o)}{\beta(x_o)} \right)$$

Equation [3-15] indicates that the normalised dynamic term of the PPG signal is influenced by the ratio of DC components originating from direct-coupled light and blood volume of non-pulsatile tissue. This ratio varies as a function of light source position, $x$.

As the light source moves away from alignment, the light path between the optoelectronics increases and this can lengthen the exposure of illumination to attenuation. In addition, diffusion from non-pulsatile tissue components reduces as the offset increases due to the increased distance between the focused area of illumination from the photodetector.

It is hard to predict the effect of direct-coupled light based solely on light source position because this effect is influenced by the tissue geometry and anatomical structure of the vasculature. However, maximum direct coupling is expected when the optoelectronics are aligned and when there are gaps where the light can traverse.
3.3.2 Applicability

The developed heuristic non-contact PPG model is supported by fundamental assumptions:

1. Measured intensity can be separated into components originating from various optical paths, and by changing the optical coupling efficiency, distinct paths can be distinguished.

2. The arterial pulsations in non-contact PPG may be modelled as a change in optical path length that modulates the coupling of light. This pulsatile component is independent of light source intensity and is a function of light source positioning with respect to the photodetector.

While core assumptions made are similar to those used by other conventional PPG models, the representations have been modified to include parameters that are significant in non-contact PPG applications. This heuristic model may be used to describe the PPG signal, but it does not determine the physical interpretation of the signal nor explain the incidence of light-tissue interactions.

3.4 Summary

A brief introduction was provided on the concept of modelling, and this was followed by descriptions of two methods that can be used to model light-matter interactions. A simple Beer-Lambert model of light-tissue interactions was expressed. It was shown that the ratio of pulsatile to non-pulsatile tissue component can provide a quantity that is proportional to changes in path length caused by blood volume change.

The other technique that can be used to model light-tissue interaction is diffusion theory. This technique can produce a more generalised model, but since the work in this thesis is aimed only as an empirical study, modelling with this technique was not attempted.
This chapter then proceeds to introduce a generalised heuristic PPG model, which separates measured intensity into components from different optical paths. This model enables the complex light interactions to be expressed in a general manner. The model was then expanded to include measurements obtained from non-contact PPG.

The primary assumption of non-contact PPG remains the same, that the received intensity can be separated into components originating from different optical paths. Measured light intensity of this heuristic model took into consideration the influence of light source positioning. The heuristic model of non-contact PPG was expanded into a more detailed arterial model.

The arterial model yields a dynamic signal that is independent of light source intensity and is expressed in the ratio of pulsatile to non-pulsatile tissue components to generate a term that is proportional to changes in path length. A first order representation of the normalised model as a function of light source position indicates that the dynamic signal component is influenced by direct-coupled light and non-pulsatile component. The heuristic models that were developed are supported by the assumption that measured light intensity can be decomposed into components from different optical paths.
CHAPTER FOUR

4 EXPERIMENTAL INVESTIGATION
4.1 Introduction

A general heuristic expression relating the effects of light-tissue interactions in non-contact photoplethsmography was developed in the previous chapter. This expression was further expanded to describe separately an arterial model and a venous model.

In this chapter, various experiments are performed to verify the predictions and assumptions made by the heuristic models and to evaluate the feasibility of the proposed non-contact PPG system. The first part of this chapter is an investigative study into the effects of direct coupling on PPG signals. Different situations are designed to examine how direct-coupled light can affect the detection of PPG signals in non-contact measurements. A prediction of normalised AC and DC signal, based on the heuristic arterial model, is presented.

The first section of this segment explores the significance of light source positioning and tissue geometry in non-contact PPG. Three different tissue geometry settings, i.e. gripped fingers, fingers with inter-digit gaps having direct exposure to direct coupling, and fingers with inter-digit gaps having indirect exposure to direct coupling, are created to demonstrate the various affects of direct coupling on PPG signal.

In the first setting, the fingers are gripped closely together to avoid any inter-digit gaps in order to minimise the detection of direct-coupled light. Meanwhile, inter-digit gaps are deliberately devised into the other two settings to generate direct coupling. In order to cater for the worst-case scenario, where detected signals are mostly direct-coupled light, the PIN photodiode in the second setting will be directly exposed to the light source. In the third setting, direct coupling is detected without having to expose the PIN photodiode directly.

In each of these three geometry settings, light source positioning will be varied by scanning the transmitter across the tissue of the fingers. The stationary photodetector will measure the illumination at different pre-determined positions along the tissue axis of the fingers. The different protocols in this section will present the different
effects that direct coupling have on PPG signals and to verify the validity of the heuristic arterial model of non-contact PPG system.

In the second section, venous occlusion plethysmography will be measured in three different settings: using a conventional PPG probe on the index finger; using non-contact PPG probe on closely gripped fingers; and using a non-contact probe on fingers with inter-digit gaps that fully expose the PIN photodiode. The intention of this section is to investigate how direct coupling affects the rate of change in the venous occlusion response and to validate the heuristic venous model.

The second segment in this chapter simultaneously measures the arterial PPG signal using the contact and non-contact probe. The agreement and repeatability between the positions of peak amplitude between the two PPG signals will be analysed. Venous occlusion plethysmography (VOP) performed in the third segment will also be measured concurrently using both contact and non-contact PPG probes. Venous occlusion curves, as well as VOP parameters including venous capacitance, arterial inflow and venous outflow, will be compared between the two acquired signals.

The second and third segments aim to examine the feasibility of non-contact PPG in both arterial and venous context respectively, by comparing the acquired signals from this method against that obtained simultaneously from conventional methods. Results presented in all the experiments are representative rather than exhaustive, their purpose being to demonstrate the practicability and feasibility of a non-contact PPG system.

4.2 Experimental Set-Up

All the experiments in this chapter were conducted in the same setting within the laboratory at the Photonics and Health Technology Research Group at Loughborough University. The same hardware equipment and software packages are used throughout this work.
4.2.1 Hardware

Non-contact PPG signals will be recorded using a DISCO4 platform manufactured by Dialog Devices, via a custom designed non-contact PPG probe as described in section 2.3.2b. The DISCO4 platform can support up to four light sources of different wavelengths, with each channel producing a raw DC and a processed AC output. These output signals are then interfaced to the computer via a USB-based data acquisition device (DAQ) for further processing.

Besides being used as an acquisition device, LabJack U12, supplied by LabJack Corporation, can also be used as an analog-to-digital converter. This device can support four differential analog input channels, with ±10 Volt input range at 12-bit resolution and an input bias current of ±90 µA. In addition, it has twenty digital input/output channels that can be configured individually, with four of them accessible through built-in screw terminals and the rest through the DB-25 connector. LabJack can thus support two of the four available light source channels of the DISCO4 platform as this device has only four differential analogue input channels.

Both contact and non-contact probes need to be connected to the DISCO4 platform in order to conduct direct comparison between the PPG signals. As the LabJack can only support two of the four light source channels, only the infrared light in both the contact and non-contact probes will be used. Only light in the infrared region was used because it has higher penetration into tissue and since infrared light is better absorbed by oxyhaemoglobin, it will provide better indication of changes in blood volume in the tissue.

4.2.2 Software

The digital data from LabJack was transferred to the computer (PC) via the Universal Serial Bus (USB) port using control software written in LabView (Laboratory Virtual Instruments Engineering Workbench), which is a graphical programming language distributed by National Instruments. The software allowed the control of LED
intensities, gain settings, sampling rates and channels configurations, displaying real-time PPG waveforms on the PC, and an option to store data on a selected location in the hard drive in the preferred prefix, as shown in Figure 4-1.

![Image of a control program for DISCO4 platform]

**Figure 4-1  Control Program for DISCO4 platform**

The stored data are imported into and analysed with Matlab, a high-level technical computing language released by The MathWorks Inc, Boston, USA, for off-line processing. The Matlab source code can be found in Appendix I. The DC signal is filtered with a moving average algorithm, which is the premier filter for time domain encoded signals to reduce random white noise while retaining a sharp step response. The noise found in the PPG signal, e.g. from movement artefact, is random; hence each sample point can be as noisy as its neighbouring points. A moving average filter is most suitable to reduce this type of noise as it treats the entire input sample equally without preference to any one of the input points, thus providing the lowest noise filter throughout the sample. The AC signal is processed by taking the ratio of the pulsatile signal to non-pulsatile signal; hence, it is a ratio that is proportional to changes in path length.
4.2.3 Choice of Anatomy

Non-contact photoplethysmography can ultimately be used to measure tissue perfusion from virtually any part of the body. However, in this initial stage of the feasibility study, measurement can only be recorded from the periphery where the tissue is relatively thin and transparent. Moreover, the anatomy of choice for this study will need to accommodate both the contact and non-contact probe.

The preferred site of measurement is the fingers, as it is simple and easy to attach both the contact and non-contact probe, and strong signals are usually found at this site. Research has shown that the highest amplitude of PPG signal, in descending order, is obtained from fingertip, cheek, anterior surfaces of lower leg, thumb and forearm\textsuperscript{[32]}. Tissue geometry of the fingers can be replicated to create inter-digit gaps that will generate direct-coupled light. As the bones of the fingers permit great mobility and flexibility, they are capable of a considerable range of movements and this enables the fingers to be shifted and repositioned without difficulty according to the requirement of the study. In addition, fingers have a rich supply of blood that originates from the ulnar and radial arteries, which form the superficial and deep palmar artery arches.

Another possible location of measurement is the foot, which is quite commonly used in PPG assessments. The foot forms a basis of support for the body, hence it is more solidly built and its component parts are less flexible than the hand. The phalanges of the foot are shorter and have very limited degree of mobility; thus, they are not as flexible as the hand and are harder to grip. The rigidity and lack of control in the digits make it harder to generate inter-digit gaps and therefore, the foot is considered unsuitable for the purpose of this study.

The earlobe is not suitable for the purpose of this experiment because the small tissue area of the earlobe will make it difficult to obtain simultaneous measurements from both contact and non-contact probes. The narrowness of the area will also limit the investigation of light source positioning. In addition, it will be impossible to change the tissue geometry of the lobe to generate direct-coupled light.
4.3 Significance of Light Source Positioning and Tissue Geometry on Non-Contact Transmission PPG

As discussed in previous chapters, the detection of direct-coupled light is potentially unavoidable in non-contact transmission PPG due to the separation of optoelectronics from skin tissue. The occurrence of this could be critical to the measured PPG signal. The possibility of detecting direct-coupled light can be influenced by light source position and the tissue geometry at the measuring site. Hence, experiments are conducted to investigate the effects of direct coupling have on non-contact PPG signals and to verify the validity of the heuristic models in the previous chapter.

The experiments in this section are designed to observe the influence of direct-coupled light on PPG signals, based on the various light source positions and tissue geometry using a non-contact transmission probe. This study will be divided into two protocols. In the first protocol, experiments are performed with fingers tightly gripped together in order to minimise the potential occurrence of direct coupling. In the second protocol, experiments will be conducted on fingers with inter-digit gaps in order to create the effects of direct coupling.

In each protocol, the light source was scanned across the finger bed from left to right, with each position along the axis separated at an interval of 5mm and each was recorded for the duration of one minute. The spatial distribution of light is adequately sampled for the purpose of this research. The area of illumination of the non-contact probe averages out the contours in the fingers. However, if a point source is being used, the sampling of spatial light distribution becomes crucial because the measured signal will be influenced by the structure of the fingers.

The light source was placed at various offset positions, \( x \), so that effects of direct coupling for different light paths could be observed. Negative and positive axes indicate the positions of the light source on the left and right of the photodetector respectively. In order to ensure consistent positioning of the light source, locations were marked prior to the start of the experiment, across 20mm to the left and 20mm to the right of the origin, respectively.
For each protocol, measurements were performed over a 5-day period on a female subject, 24 years of age and healthy, with no known history of vascular disease. The duration of the experiment was decided so that the whole experiment would be completed within one week in order to minimise physiology and ambient changes between the measurements.

\[ -\text{ve } x \quad x_0 \quad +\text{ve } x \]

\[ \text{Light Source} \]

\[ \text{Finger Bed} \]

\[ \text{Photodetector} \]

\[ (a) \quad (b) \]

**Figure 4-2 Schematic Illustration of the Principles of Non-Contact Transmission PPG with (a) minimised and (b) maximised presence of direct coupling**

The subject was asked to sit upright on a chair, relax and breathe normally whilst keeping the left hand as still as possible on the platform of the non-contact transmission PPG prototype during recording. The light source and photodetector were positioned approximately 80mm apart, one on each side of the vasculature. When the lateral positioning distance is reduced, subjects with larger hands have difficulty in positioning the fingers on the platform of the non-contact PPG platform. On the other hand, at a larger lateral positioning, there are situations when the illumination from the light source is not strong enough to reach the photodetector. As a result, the lateral positioning is fixed because a high quality PPG signal could be recorded at this distance. With the photodetector in a fixed location, the alignment between the light source and photodetector was referred to as the origin or centre, \( x_0 \).
The light source position was shifted across the anatomy along the x-axis, as shown in Figure 4-2.

4.3.1 Model Prediction

Values for the coupling coefficients were predicted from the measured PPG signals obtained in the experiment. The average values of these coefficients were obtained from solving the simultaneous equations of the heuristic model and are recorded in Table 4-1. It is not possible to separately identify the coupling coefficients that are coupled via direct coupling and non-pulsatile tissue, and hence, they will be aggregated together. The Beer-Lambert law is a simple first order model used to aid the understanding of transmission of light through tissue, it does not consider optical effects such as diffusion and scattering in tissue. It is believed that these optical effects would influence the path length of light and hence, affect the calculated coefficient values.

Figure 4-3 shows the normalised AC and DC signal as predicted by the heuristic model. The normalised DC signal has a symmetrical bell shaped graph that peaks at the origin. The normalised AC PPG signal is also a smooth symmetrical shaped graph that is centralised at alignment. As the light source shifts away from the alignment position, the increased separation between the optoelectronics decreases the amount of illumination that is detected by the photodetector.
<table>
<thead>
<tr>
<th>Light Source Position (mm)</th>
<th>$\alpha + \beta$</th>
<th>$\gamma$</th>
</tr>
</thead>
<tbody>
<tr>
<td>-20</td>
<td>0.447</td>
<td>1.764</td>
</tr>
<tr>
<td>-15</td>
<td>0.657</td>
<td>1.304</td>
</tr>
<tr>
<td>-10</td>
<td>0.837</td>
<td>1.109</td>
</tr>
<tr>
<td>-5</td>
<td>0.957</td>
<td>1.020</td>
</tr>
<tr>
<td>0</td>
<td>0.999</td>
<td>0.987</td>
</tr>
<tr>
<td>5</td>
<td>0.957</td>
<td>1.001</td>
</tr>
<tr>
<td>10</td>
<td>0.837</td>
<td>1.078</td>
</tr>
<tr>
<td>15</td>
<td>0.657</td>
<td>1.290</td>
</tr>
<tr>
<td>20</td>
<td>0.447</td>
<td>1.863</td>
</tr>
</tbody>
</table>

*Table 4-1 Values of Coupling Coefficients used in the Prediction Model*

*Figure 4-3 Prediction Model of Normalised PPG Signal*
4.3.2 Measurements on Gripped Fingers

In the protocol performed on gripped fingers, ten sets of measurements were recorded successfully over a 3-day period using the non-contact transmission system. The signals were recorded between the range of 20mm to the left to 20mm to the right of alignment. However, only nine measurements were analysed, as one of the source-positioning recordings was missing in the discarded data set. Both the DC and AC signals were normalised to their respective signal components that were measured during alignment.

<table>
<thead>
<tr>
<th>Light Source Position (mm)</th>
<th>Mean DC Voltage (V)</th>
<th>Mean AC Voltage (V) at 46dB gain</th>
<th>DC normalised (a.u.)</th>
<th>AC normalised (a.u.)</th>
</tr>
</thead>
<tbody>
<tr>
<td>-20</td>
<td>0.74</td>
<td>1.68</td>
<td>0.292 ± 0.025</td>
<td>0.819 ± 0.168</td>
</tr>
<tr>
<td>-15</td>
<td>1.30</td>
<td>1.77</td>
<td>0.515 ± 0.049</td>
<td>0.861 ± 0.133</td>
</tr>
<tr>
<td>-10</td>
<td>2.03</td>
<td>1.88</td>
<td>0.804 ± 0.057</td>
<td>0.916 ± 0.109</td>
</tr>
<tr>
<td>-5</td>
<td>2.45</td>
<td>1.87</td>
<td>0.972 ± 0.046</td>
<td>0.914 ± 0.191</td>
</tr>
<tr>
<td>0</td>
<td>2.53</td>
<td>2.05</td>
<td>1.000 ± 0.000</td>
<td>1.000 ± 0.000</td>
</tr>
<tr>
<td>5</td>
<td>2.09</td>
<td>2.28</td>
<td>0.825 ± 0.061</td>
<td>1.113 ± 0.153</td>
</tr>
<tr>
<td>10</td>
<td>1.60</td>
<td>2.20</td>
<td>0.631 ± 0.057</td>
<td>1.073 ± 0.220</td>
</tr>
<tr>
<td>15</td>
<td>1.12</td>
<td>2.47</td>
<td>0.441 ± 0.053</td>
<td>1.205 ± 0.186</td>
</tr>
<tr>
<td>20</td>
<td>0.68</td>
<td>2.01</td>
<td>0.267 ± 0.034</td>
<td>0.980 ± 0.104</td>
</tr>
</tbody>
</table>

Table 4-2 Mean Relative Amplitude of PPG Signals for 9 Measurements Recorded Using Non-Contact Transmission PPG

Table 4-2 shows the mean DC and AC voltages, as well as the mean DC and AC normalised (± standard deviation) amplitude of PPG signals measured at each position along the fingers. In this table, the mean AC voltage is amplified at a gain of 46dB. The measured AC normalised components, in this table and throughout the experiment, are the normalised dynamic term of PPG signals that are influenced by the ratio of DC components. Figure 4-4 shows plots of the relative mean peak-to-peak amplitude for both the DC and AC components at the various positions along the vasculature.
The change in the shape of the measured normalised DC signal is in agreement with the predicted model, where maximum illumination is detected when the light source and the photodetector are in alignment. However, there is a considerable difference in the predicted and the measured normalised AC signal. The normalised AC signal exhibited higher amplitudes on the right axis in the measured AC plot, which could be caused by the detection of direct coupling.

The relative mean amplitude of the DC signal in gripped fingers has a bell-shaped graph that peaked at the origin. As the offset increased between the optoelectronics, the DC signal decreased gradually. This can be explained by the reduced illumination as the distance lengthens between the light source and the photodetector. The standard deviation obtained at each position was small, indicating that consistent illumination was measured across the different sets of measurements.

The DC signals on the left of the photodetector exhibited a comparatively higher measure of mean amplitude, especially at the offset position of 5mm. At this
particular position, the difference between the measured mean amplitude with the maximum amplitude detected at the origin is only 2%. This higher amplitude measurement can be an indication of the presence of direct coupling. This could be a result of the anatomical structure of the fingers, as no matter how tightly they are gripped, there are bound to be small inter-digit gaps caused by the geometry in the fingers. We can support this by examining the normalised mean peak-to-peak amplitude of the AC components.

The heuristic model in Equation [3-14] suggests that the normalised mean peak-to-peak AC amplitude would decrease when direct-coupled light in relation to tissue DC component increases. This decrease in amplitude can be observed in the AC plot of Figure 4-4. The normalised mean peak-to-peak AC amplitude is expected to decrease when there is either an increase in direct-coupled light or a reduction in illumination due to non-pulsatile tissue. The tissue area on the fingers is very similar, therefore it is believed that the decrease in amplitude is a result of the detection of direct coupling. Nevertheless, the amount of direct coupling detected at this position was not significant enough to reduce the SNR of the AC signals, as shown in Figure 4-5.

On the right axis of photodetector, the mean DC amplitude can be seen to decrease gradually with increasing light source positions. There was no obvious sign of direct coupling on this axis of measurement. The normalised mean peak-to-peak AC signal seems to reflect this deduction, as higher amplitudes were measured compared to the left axis.

The heuristic model correctly predicts the change in normalised DC signal but does not accurately predict the change in normalised AC signal. The reason for this is that the first-order Beer-Lambert law is only limited to the prediction of light transmission and does not take into account the scattering and diffusion of light in the tissue. Diffusion theory or Monte Carlo simulation can be used in the future to obtain a better and more detailed analysis of light-tissue interactions. From the experiment, it can be concluded that the quality of measured PPG signals using non-contact transmission PPG is not compromised if the extent of direct-coupled light is kept to a minimum.
4.3.3 Measurements on Fingers with Inter-Digit Gaps

Two sets of experiments were performed to measure PPG signals on fingers with inter-digit gaps. The first set of measurements were conducted with the inter-digit gap between the index and middle fingers made large enough to expose the PIN photodiode in the receiver to the light source when aligned. This is to cater for the worst-case scenario where the photodetector detects mainly direct-coupled light from the light source. The second set of measurements avoids the direct exposure of the PIN photodiode by shifting the fingers to the right, at the same time maintaining the inter-digit gaps, so that the middle finger physically blocks the direct illumination of light during alignment.
Each set of experiments were conducted in a day and on both occasions, only four measurements were recorded. It is comparatively harder to record PPG signals from fingers with inter-digit gaps because of the increased detection of movement artefacts, ambient artefacts and direct coupling.

4.3.3a Direct Exposure of Direct Coupling to PIN Photodiode

Figure 4-6 illustrates the relative mean DC amplitude that was measured on fingers with inter-digit gaps that fully exposed the PIN photodiode to the light source. The measured mean DC amplitude has a different response compared to DC signals measured by the other two protocols, as illustrated in Figure 4-4 and Figure 4-14. The maximum DC amplitude was detected when the light source was positioned at the 5mm offsets on both the left and right axes. Sharp decreases in amplitude were observed when the light source was shifted away from these two positions.

![DC Signals](image)

**Figure 4-6 Mean Amplitude on Fingers with Direct PIN Photodiode Exposure Using Non-Contact Transmission PPG**
It can be argued that the massive increase in amplitude towards alignment is a result of direct coupling since the PIN photodiode is directly exposed to the illumination from the light source. As direct coupling is a measure of the DC component, the extent of this effect is clearly portrayed in Figure 4-6. As the light source was positioned away from alignment, the amount of direct coupling reduced substantially, hence the great drop in signal amplitude. Nevertheless, we would have expected the maximum illumination to be detected during alignment and not at the offset positions. This situation, however, can be explained by tissue geometry of the fingers.

During the experiment, the gap between the index and middle fingers was carefully constructed so that the PIN photodiode would be exposed, but at the same time, would not saturate the photodetector. Some tissue would still reside in between the non-contact transmission probes, as illustrated in Figure 4-7.

![Figure 4-7](image)

*Figure 4-7 Illumination of Fully Exposed Photodetector of the Non-Contact Transmission PPG Probe at Alignment*

When the optoelectronics are aligned, most of the light intensity is likely to be illuminated directly on the photodetector rather than reflected from the sides of the reflector probe. In addition, light reflected off the sides of the fingers could be altered from the original light path and fall into region outside the photodetector probe. Illumination collected by the reflector probe would most probably have originated from the vasculature.
When the light source was positioned at certain offset positions, the photodetector could be more exposed to illumination directly from the light source if the inter-digit gap is large enough, as shown in Figure 4-8. The receiver could detect direct-coupled light through the reflector probe and caused multiple reflections within the probe before the illumination reached the PIN photodiode.

*Figure 4-8  Illumination of Fully Exposed Photodetector of the Non-Contact Transmission PPG Probe at an Offset*

As intensity of direct-coupled light is stronger than light from tissue, the measured DC signal would be much larger. From Figure 4-6, the offset positions in this experiment occurred at both sides of the alignment, at 5mm to the left and 5mm to the right. In addition, as the photodetector is out of alignment, light originating from the tissue might not be collected by the photodetector.

Mean DC amplitudes at other offset positions are nearly twice as low in amplitude compared to that measured on gripped fingers. In the experiment with gripped fingers, light illuminated on the fingers would be diffused to tissue areas in the adjacent region. Consequently, the reflector probe would also gather light from the diffused tissue area. However, when inter-digit gaps are present, light could not be diffused to surrounding tissue areas. Hence, measurement of DC amplitude could have stemmed from a lower tissue area, resulting in a reduced DC amplitude.
Figure 4-9  Sample of AC Signal each Offset Position as Measured by Non-Contact Transmission PPG

Figure 4-10 Normalised Autocorrelation for 10s of AC Signals at 4 Source Positions as Measured by Non-Contact Transmission PPG
Figure 4-9 shows a sample of AC signal recorded from each of the measurement positions. Amplitude of the AC signal is small (±2V or less) at offside positions: left 20mm, left 15mm, left 10mm, right 15mm and right 20mm. A PPG shaped waveform can be recognised from these positions. The DISCO4 platform could not detect arterial PPG signals when the light source was positioned at 5mm offset to the left, during alignment and at both 5mm and 10mm offsets to the right of photodetector. Amplitude of signal at these positions are comparatively higher. Figure 4-10 shows the autocorrelation plot of the AC signals at each of these locations and they do not indicate any obvious correlation in the PPG signals at each of these positions.

When a large extent of direct-coupled light was detected, at offset positions: left 5mm, alignment, right 5mm and right 10mm, the AC signals measured resembled some random signals. When these signals were passed through the autocorrelation function, no strong periodic repetitions could be detected, but rather, very weak fluctuations around zero amplitude were observed. These fluctuations could indicate that some very weak PPG signals might reside within the random signals. This is expected because light was transmitted through some tissue area, but the amount of measured intensity due to this is very small, as shown in Figure 4-11. The direct-coupled light would have detected mostly random skin surface movements, hence generating signals with high degree of motion artefacts.

At the other positions where the mean DC amplitudes were low, strong arterial PPG signals were detected and this is shown in the autocorrelation plot in Figure 4-12. As the light source was shifted away from the direct illumination path, the proportion of detected light resulting from direct coupling reduces, and hence signals from the tissue area increase to a level where it is observable again, as seen in Figure 4-13.

The detection of motion artefacts is a dilemma faced in photoplethysmography and pulse oximetry applications. On-going research has been conducted to resolve the problem posed by motion artefacts, but thus far they have only managed to minimise the effect rather than completely resolve it\cite{114,121}.
Figure 4-11  Reduced Proportion of Arterial Blood Due to Direct Coupling

Figure 4-12  Autocorrelation for 10s of AC Signals for Remaining Light Source Positions as Measured by Non-Contact Transmission PPG
In this protocol, it was observed that in the presence of direct coupling, detection of AC signals was not possible because the proportion of light due to direct illumination was too strong. In addition, direct-coupled light could pick up random tissue movements on skin surface and these motion artefacts would corrupt the detected AC signals. However, when the amount of direct-coupled light reduces and illumination resulting from tissue increases, AC signals was detected again. Hence, it was demonstrated that in the strong presence of direct coupling, the measured AC signals using non-contact transmission PPG are highly influenced by motion artefacts.

4.3.3b Indirect Exposure of Direct Coupling

The second set of protocols was performed with inter-digit gaps, but direct exposure of the PIN photodiode was avoided. The middle finger was positioned between the non-contact transmission probes to obstruct any direct illumination, while inter-digit gaps still existed between the index-middle fingers and ring-middle fingers. Figure
4-14 shows the mean DC amplitudes that were measured by the photodetector. The maximal intensity was not detected during alignment but at the 5mm offset to the left of the photodetector. In addition, the mean DC amplitudes on the left axis showed greater variations in measured amplitudes. These variations could again indicate the presence of direct coupling due to inter-digit gaps.

![DC Signals Graph](image)

*Figure 4-14  Mean Amplitude Measured without Direct Exposure to PIN Photodiode as Measured by Non-Contact Transmission PPG*

Table 4-3 shows the periodicity of the autocorrelation plot on the AC components from the four data sets collected using this protocol. No consistency in the pattern or trend can be observed. The detailed autocorrelation plot can be found in Appendix J. When the experiments were performed, the middle finger always resided in between the non-contact transmission PPG probe, but the magnitudes of the inter-digit gaps were not specified. This could be the reason behind the inconsistency in the measured AC signals. From the table, the trend of how the AC signals were affected by direct coupling cannot be deduced, but it was shown that in the presence of direct-coupled light through inter-digit gaps, the quality of the measured AC signals is equivocal.
The AC components of the non-contact transmission PPG signals can be recovered when direct-coupled light is inadequate to overshadow the illumination resulting from tissue perfusion. The signal-to-noise ratio of the AC signal depends on the tissue geometry and the amount of direct coupling that was present during the measurement. When there was no direct illumination from the light source to the photodetector, the chances of retrieving the PPG signals are high. Therefore, to ensure the detection of good signal-to-noise ratio of PPG signals, inter-digit gaps have to be avoided.

In this experiment, the auto-correlation function was used to check for periodicity of the AC signals to identify if PPG signals existed in the measured waveform. The non-contact probe will have to be developed further before it is acceptable for any PPG applications. Issues such as periodicity and direct coupling will have to be addressed, yielding a high quality PPG signal that is comparable to signals measured from a conventional PPG probe.

### 4.4 Significance of Tissue Geometry on Non-Contact Reflection Mode PPG

An experiment was designed to investigate the effect of direct coupling on PPG signal by positioning the reflection probe across various preset positions along the hand and by changing the anatomical positioning of the fingers. This study is divided into two
protoclos: in the first protocol, experiments were performed on fingers that are tightly
gripped together, as shown in Figure 4-15a; and in the second protocol, experiments
were carried out on fingers with inter-digit gaps, as shown in Figure 4-15b. The light
source was scanned across the finger bed from left to right, with each position along
the axis separated at 5mm intervals, the PPG signal was recorded from each position
for one minute.

![Non-contact reflection probe](image)

**Figure 4-15  Schematic Illustration of the Non-Contact Reflection Mode PPG with (a)
gripped fingers and (b) fingers with inter-digit gaps**

For each protocol sets, measurements were performed on a healthy, 28 years of age,
female subject in a room with a warm ambient temperature. The subject was asked to
sit upright on a chair, relax and breathe normally, whilst keeping the left hand palm up
facing the non-contact reflection probe on the PPG platform as still as possible during
recording to minimise movement artefact. The light source was positioned
approximately 50mm above the hand. The origin or centre, \( x_0 \), refers to the position
in the middle of the hand above the middle finger, and \( x \) refers to the various offset
positions to the left and right of origin.
4.4.1 Measurements on Gripped Fingers

In the protocol on gripped fingers, ten successful measurements were recorded over a 3-day period. The signals were recorded between the range of 20mm to the left and 20mm to the right of the origin $x_0$. Both the DC and the AC signals were normalised to their respective signal components that were measured at the origin.

Table 4-4 shows the mean (± standard deviation) amplitude of the PPG signals measured at each position along the axis. Figure 4-16 shows the relative mean peak-to-peak amplitude for both the DC and the AC components that were recorded for each position.

<table>
<thead>
<tr>
<th>Light Source Position (mm)</th>
<th>Mean DC Voltage (V)</th>
<th>Mean AC Voltage (V) at 46dB gain</th>
<th>DC normalised (a.u.)</th>
<th>AC normalised (a.u.)</th>
</tr>
</thead>
<tbody>
<tr>
<td>-20</td>
<td>2.43</td>
<td>0.96</td>
<td>0.758 ± 0.104</td>
<td>1.033 ± 0.340</td>
</tr>
<tr>
<td>-15</td>
<td>2.70</td>
<td>1.06</td>
<td>0.845 ± 0.105</td>
<td>1.051 ± 0.342</td>
</tr>
<tr>
<td>-10</td>
<td>2.95</td>
<td>1.16</td>
<td>0.922 ± 0.076</td>
<td>1.194 ± 0.333</td>
</tr>
<tr>
<td>-5</td>
<td>3.15</td>
<td>1.24</td>
<td>0.983 ± 0.055</td>
<td>1.011 ± 0.163</td>
</tr>
<tr>
<td>0</td>
<td>3.20</td>
<td>1.26</td>
<td>1.000 ± 0.000</td>
<td>1.000 ± 0.000</td>
</tr>
<tr>
<td>5</td>
<td>3.21</td>
<td>1.26</td>
<td>1.003 ± 0.062</td>
<td>1.163 ± 0.337</td>
</tr>
<tr>
<td>10</td>
<td>3.03</td>
<td>1.19</td>
<td>0.947 ± 0.097</td>
<td>1.205 ± 0.292</td>
</tr>
<tr>
<td>15</td>
<td>2.78</td>
<td>1.09</td>
<td>0.869 ± 0.082</td>
<td>1.130 ± 0.254</td>
</tr>
<tr>
<td>20</td>
<td>2.37</td>
<td>0.93</td>
<td>0.741 ± 0.105</td>
<td>1.171 ± 0.209</td>
</tr>
</tbody>
</table>

*Table 4-4 Mean Relative Amplitude of PPG Signals for 10 Measurements asMeasured by Non-Contact Reflection Probe*

The relative mean amplitude for the DC component has a smooth curve that peaks around the origin, and the mean amplitude is relatively symmetrical around the origin, reflecting the general shape of the fingers. The standard deviation from each position was small and this implied that consistent measurements were obtained across the 10 sets of PPG recordings. The DC amplitude dips at the two edges because as the
reflection probe shifted towards the thumb and the little finger, the amount of tissue in
the field of view of the probe decreases.

![DC Signal](image1)

![AC Signal](image2)

**Figure 4-16 Mean Relative Amplitudes Measured on Gripped Fingers Using Non-Contact Reflection PPG**

The normalised mean peak-to-peak AC component showed some fluctuations in
amplitude, this may be caused by the contours of the finger. As the light sources in the
non-contact reflection probe has a narrower viewing angle, its illumination would not
be as diffused as illumination from a non-contact transmission probe. Hence, it is
more sensitive to the changes in contours of the fingers.

The relative amplitude of the AC signal peaked at the 10mm offset positions on both
the left and the right axis, which also happens to be the edge separating the two
adjacent fingers. In these positions, the photodetector was positioned on the line
between the two fingers and this meant that most of the light sources were positioned
directly on top of the fingers, and this would improve the degree of light reflected
back to the probe. As a result, the observed increase in signal amplitude is likely to be
due to the increase in reflection from the fingers and the decrease in illumination loss
due to the inter-digit gaps.
When the non-contact reflection probe is at other offset positions, some of the light is illuminated on the tissue where the fingers are separated. There are bound to be small inter-digit gaps caused by the geometry of the fingers no matter how tightly the fingers are gripped, and hence, some of the light illumination could be lost. In addition, the slope between two fingers at separation reflects light differently depending on the angle of light interaction, and hence light might be reflected outside the field of view of the photodetector. The loss in light illumination from these circumstances is reflected in the relative amplitude being lower at these offset positions.

Figure 4-17 illustrates a 10 second sample of the AC component of the non-contact reflection PPG signal at each offset position. A PPG shaped waveform can be recognised from all these positions. Figure 4-18 and Figure 4-19 show the autocorrelation plot for the AC signals and it is obvious from the figures that a strong correlation exists in the AC PPG signals at each position.
Figure 4-18 Normalised Autocorrelation for 10s of AC Signals at the Left Offset Positions on Gripped Fingers Using Non-Contact Reflection PPG

Figure 4-19 Normalised Autocorrelation for 10s of AC Signals at Origin and Right Offset Positions on Gripped Fingers Using Non-Contact Reflection PPG
4.4.2 Measurements on Fingers with Inter-Digit Gaps

The protocol for the experiment of measurements on fingers with inter-digit gaps was the same as that conducted with gripped fingers. The main purpose for this experiment is to investigate the effect that changing tissue geometry has on PPG signal. In order to cater for similar tissue geometry throughout the different sets of measurement, the gaps in the fingers were fixed to be about the size of the PIN photodiode.

The experiment was conducted in a day and five measurements were successfully recorded. Table 4-5 shows the mean amplitude (± standard deviation), whilst Figure 4-20 shows the plot of the relative mean amplitude, of both the AC and the DC components of the PPG signal at each of the offset positions.

<table>
<thead>
<tr>
<th>Light Source Position (mm)</th>
<th>Mean DC Voltage (V)</th>
<th>Mean AC Voltage (V) at 46dB gain</th>
<th>DC normalised (a.u.)</th>
<th>AC normalised (a.u.)</th>
</tr>
</thead>
<tbody>
<tr>
<td>-20</td>
<td>1.66</td>
<td>1.78</td>
<td>0.804 ± 0.156</td>
<td>1.170 ± 0.358</td>
</tr>
<tr>
<td>-15</td>
<td>2.05</td>
<td>1.79</td>
<td>0.988 ± 0.040</td>
<td>1.175 ± 0.247</td>
</tr>
<tr>
<td>-10</td>
<td>2.06</td>
<td>1.59</td>
<td>1.017 ± 0.054</td>
<td>1.048 ± 0.178</td>
</tr>
<tr>
<td>-5</td>
<td>2.03</td>
<td>1.58</td>
<td>0.980 ± 0.065</td>
<td>1.041 ± 0.225</td>
</tr>
<tr>
<td>0</td>
<td>2.07</td>
<td>1.52</td>
<td>1.000 ± 0.000</td>
<td>1.000 ± 0.000</td>
</tr>
<tr>
<td>5</td>
<td>2.09</td>
<td>1.55</td>
<td>1.008 ± 0.113</td>
<td>1.019 ± 0.264</td>
</tr>
<tr>
<td>10</td>
<td>2.04</td>
<td>1.54</td>
<td>0.985 ± 0.110</td>
<td>1.013 ± 0.237</td>
</tr>
<tr>
<td>15</td>
<td>2.03</td>
<td>1.60</td>
<td>0.982 ± 0.073</td>
<td>1.054 ± 0.163</td>
</tr>
<tr>
<td>20</td>
<td>1.89</td>
<td>1.60</td>
<td>0.913 ± 0.078</td>
<td>1.053 ± 0.223</td>
</tr>
</tbody>
</table>

Table 4-5 Mean Relative Amplitude of Non-Contact Reflection Mode PPG Signals for 5 Measurements

Except for the two positions on the edge, each of the other offset positions have recorded relatively similar measures of mean DC amplitude, an effect from the spread of fingers and inter-digit gaps. The standard deviation is relatively small at each
position and this indicates that reliable measurements were recorded in the 10 sets of measurements.

![DC Signal](image)

![AC Signal](image)

**Figure 4-20  Mean Relative Amplitudes Measured on Fingers with Inter-Digit Gaps Using Non-Contact Reflection PPG**

In this experiment, the relative mean AC signal does not show fluctuations in its measurement as was observed on gripped fingers because light reflected from the skin surface and light lost through the inter-digit gaps would even out at the offset positions. Figure 4-21 shows a 10 second sample of PPG signals at each offset position that was recorded on fingers with inter-digit gaps using a non-contact reflection probe.

All of the recordings at the offset positions showed PPG shaped waveforms. Although the AC signals appeared to be noisy and not as consistent in their shape, the autocorrelation plots in Figure 4-22 and Figure 4-23 verify that a strong correlation exists in these AC signals. This experiment confirmed that good signal-to-noise ratio with strong periodicity of AC signal were recorded even when there is an inter-digit gap in the fingers. Hence, the proposal that tissue geometry at the site of measurement does not affect the PPG signal measured by the non-contact reflection probe is validated.
Figure 4-21 10s Sample of PPG Signal Recorded on Fingers with Inter-Digit Gaps at Each Offset Positions

Figure 4-22 Normalised Autocorrelation of AC Signals at the Left Offset Positions on Fingers with Inter-Digit Gaps
4.5 Comparison of Arterial Signal by Contact and Non-Contact PPG Systems

Earlier experiments have shown that satisfactory PPG signals can be measured using both the proposed transmission and reflection non-contact system, provided that direct coupling is kept to the minimum when using a non-contact transmission PPG probe. In this section, further investigations analyse the feasibility of using the proposed non-contact PPG systems through simultaneous comparison of arterial PPG signals that are obtained by contact and non-contact PPG probes.

Various studies have been conducted using arterial PPG signals to investigate physiological factors, including respiratory rate, blood flow, arterial blood pressure, and to study properties of arterial pulse by analysing the wave velocity, the pulse transit time and the pulse contour. Analyses
EXPERIMENTAL INVESTIGATION

performed in these studies were mainly based on the amplitude or peaks of the PPG signals.

This study makes a direct comparison of the simultaneous AC signals that were recorded by contact and non-contact PPG probes. Before the start of the experiment, the subject was asked to relax for 2 minutes, and then data were recorded for 3 minutes with the subject breathing at a normal rate. It is important to investigate the similarity in the qualitative measure of the PPG signal detected by the two probes. The quantitative measure of the signal amplitude is less crucial because this can be modified by adjusting the intensity and gain control of the DISCO4 platform.

For the subsequent experiments, the Bland Altman plot was employed as the statistical method to analyse the agreement between the contact and the non-contact PPG systems. The limits of agreement refer to the approximation that if the differences are normally distributed (Gaussian), 95% of the differences will lie between these limits. The 95% confidence interval (CI) refers to the precision of the limit of agreement calculated by finding the appropriate point of the \( t \) distribution with \( n-1 \) degree of freedom. This means that if the 95% CI is small, it gives confidence that there is good agreement between the two systems.

4.5.1 Non-Contact Transmission Mode

The experiment using non-contact transmission mode PPG was performed on fourteen subjects, seven male and seven female, aged between 21 to 53 (31.50±8.94) years of age, with no known vascular diseases. A disposable, adhesive, pulse oximeter sensor (Nellcor OxiMax Max-A/Max-AL, Nellcor, CA, USA) was carefully attached on the subject's left index finger and the same hand was then positioned onto the non-contact PPG platform fixed with the transmission probe, ensuring that the fingers were closely gripped together with no apparent inter-digit gaps.
Signals from Contact and Non-Contact Transmission PPG

Figure 4-24  PPG Signal Measured by Contact and Non-Contact Transmission PPG

Figure 4-24 shows a 5s recording taken from one of the female subjects during the experiment. It can be observed that arterial PPG signals measured by both contact and non-contact transmission probes have very similar profile during systole, but the non-contact PPG signal appears to be noisier at diastole.

Figure 4-25 shows a comparison of PPG peak positions measured by a conventional transmission mode PPG probe (CPPG) and the proposed transmission mode non-contact PPG probe (NTPPG) using a Bland-Altman plot\cite{126}. Owing to the 100Hz sampling rate of the DISCO4 platform, it is noted that the differences between the positions of the amplitudes have a bias of 10ms. It is observed that one of the average data points was located outside the limit of agreement. It was identified that this was caused by the detection of involuntary movement in the fingers by the non-contact transmission probe.

The mean difference between the two systems is 7.5ms with 95% confidence interval (CI) of the mean difference at 5.9ms and 9.1ms. Hence, the non-contact transmission PPG peak position, as measured by the DISCO4 platform, tends to lead the contact PPG peak position by one sampling period. This bias is a result of the time-division multiplexing scheme adopted in the DISCO4 platform, as only one light source will
be illuminated at any one sensing period. The limits of agreement at -22ms (95% CI at
-24.8ms and -19.2ms) and 7ms (95% CI 4.2ms and 9.8ms) are small enough to deduce
that the peak positions measured by non-contact PPG in this experiment are in
agreement with peak positions measured by conventional PPG.

![Graph showing peak amplitude position]

**Figure 4-25 Amplitude Position as Measured by Contact and Non-contact Transmission Mode PPG**

Figure 4-26 displays the mean differences between the two methods across the 14
subjects. The figure shows the mean differences between peak positions measured in
non-contact and contact PPG for each subject. All but one of the data sets lies
between the limit agreement of -13.3ms and 10.7ms. It was observed from these data
that the non-contact PPG signal is relatively noisy, especially in diastole, which could
be an effect from the detection of direct-coupled light.

The overall measurements projected a mean difference value of 1.3±6ms, which is
low enough to argue that there is very little difference to separate between the peak
positions measured by the two systems. This demonstrates that measurements
obtained by the two methods are highly repeatable, with the PPG peak positions
detected within one sampling period of each other.
4.5.2 Non-Contact Reflection Mode

The experiment using a non-contact PPG probe in reflection mode was conducted on twelve subjects, seven female and five male, aged between 25 to 52 (33.00±8.75) years of age, with no known vascular diseases. A disposable, adhesive, pulse oximeter sensor (Nellcor OxiMax Max-A/Max-AL, Nellcor, CA, USA) was carefully attached to the subject's left finger and the hand was carefully positioned on the non-contact reflection PPG platform. The subject was asked to grip their fingers closely together, relax and not to move their hand during the recording session.

Figure 4-27 shows a 5 second sample of PPG signals recorded simultaneously using contact and non-contact reflection probes from one of the female subjects. It is observed that the profile of the PPG signal recorded from non-contact PPG is very similar to that recorded from contact probe; from the rising peak phase, which corresponds to the systole phase in the cardiac cycle and to the trough, which
corresponds to the diastole phase, and even the dichrotic notch can be clearly seen from the non-contact PPG signal.

*Figure 4-27  Samples of PPG Signal Measured by Contact and Non-Contact Reflection Mode PPG*

*Figure 4-28  Amplitude Positions as Measured by Contact and Non-Contact Reflection Mode PPG*
Figure 4-28 shows a comparison of PPG peak positions as measured by the standard PPG contact probe (CPPG) and the non-contact reflection mode probe (NRPPG). The differences between the peak positions have a bias of 10ms, and this is due to the sampling period of the DISCO4 platform.

The mean difference between the two PPG systems is -9.7ms with 95% CI at -9.0ms and -10.4ms. The limits of agreement at 1.7ms (95% CI at 0.4ms and 2.9ms) and -21ms (95% CI at -22.3ms and -19.8ms) are small, and hence it is deduced that the peak positions as measured by contact and non-contact reflection mode PPG are in agreement with one another.

Figure 4-29 shows the plot of the mean differences of the peak positions as measured across the 14 measurements. The mean value is -1.1ms and the 95% CI is at 0.9ms and 0ms. The limit of agreement is between 14.1ms (95% CI at 17.6ms and 10.6ms) and -16.3ms (95% CI at -12.7ms and -16.3ms). All the data sets are within the limit agreement, and hence, this shows that the PPG signal measured by the two PPG systems are highly repeatable.
4.5.3 Discussion

PPG signals measured from the non-contact system, operating in both transmission and reflection mode, have similar response to the PPG signal measured from standard contact probe. Besides being noisier at the trough of the PPG pulses, arterial waveforms measured by non-contact transmission probe were very similar to those measured by the conventional method, and it is predicted that skin tissue movement during the diastole phase causes the noise in the PPG signal. Signals recorded using non-contact reflection probe are very similar to those recorded from a contact PPG probe, and the noise at the trough of the PPG pulse is much less than that observed using non-contact transmission probe.

Blood is ejected from the heart during ventricular systole and is drained during diastole. Arteries have thick elastic tissues that can expand and contract with each heartbeat, and therefore when blood is ejected during systole, the arteries expand to accommodate the increase in blood volume, and the arteries contract when blood is drained back to the heart. Hence, the skin tissue will also move in synchrony with the expansion and contraction of the arteries.

Using a conventional probe, the light source and photodetector are attached directly on the vasculature. Therefore, when there are tissue movements, the optoelectronics move in the same direction as the skin surface, hence cancelling any movement effects that have occurred. In addition, the force applied by the probe onto the vasculature might press down the skin, thus suppressing any skin surface movements as well as reducing the change in amplitude of the PPG signals\cite{41,78} that is due to movement artefact.

On the other hand, the optoelectronics in non-contact PPG are positioned away from the skin surface. The expansion and contraction of the arteries generate surface tissue movements that will alter the path length of illumination from the light source to the photodetector. During systole, expansion of the arteries causes the skin to stretch out, thus increasing the tissue area, smoothes the skin and fills up any anatomical gaps. The surface tissue then loses the internal pressure from the arteries during contraction.
of the heart. This would cause the skin to shrink at a random rate, depending on the elasticity of the arteries and the skin, as well as the surface texture of the skin. Hence, this could explain the noisier PPG signals that are detected during diastole by non-contact probe and this demonstrates the sensitivity of non-contact PPG probes in detecting surface tissue movements.

Nevertheless, it is clearly shown that the amplitudes of the non-contact PPG signal, as measured by both non-contact transmission and reflection probe, is in good agreement with that measured using a conventional contact probe. The experimental results show that the mean differences in peak positions measured by both the non-contact transmission and reflection PPG probes are very small. Most of these differences were found between the 95% confidence interval, thus indicating that the position of the PPG peaks measured by contact and non-contact probes have a high degree of agreement. Hence, non-contact PPG can be used as a substitute for conventional PPG in peak detection applications.

4.6 Evaluation of Venous Occlusion Plethysmography Measured By Contact and Non-Contact PPG System

Photoplethysmography provides a non-invasive assessment of blood volume and tissue perfusion in the tissue. The light intensity measured by the optoelectronics is comprised of two signal components; one attributable to the pulsation in the vessels caused by the heart beat (AC), and the other a quasi-static signal attributable to the absorption by tissue structures and venous blood (DC).

As the AC component oscillates with each heart cycle period, variations in signals acquired between contact and non-contact PPG systems can be easily identified. On the other hand, the slow fluctuations in DC signals do not provide any apparent measurements that can be significantly compared when measured on their own. Therefore, venous occlusion plethysmography (VOP) is performed to induce a change to the venous blood volume in order to enable a substantial comparison to be conducted between the two PPG probes.
The venous occlusion plethysmography protocol employed in this and subsequent experiments is similar to the protocol employed for the evaluation of arterial and venous function in the lower limb\cite{711}. In standard practice, VOP used in the study of forearm blood flow excludes the circulation from the hand. The reason for doing this is that blood flow in the hand is predominantly through the skin and has a high proportion of arteriovenous shunts, and thus it has a different pharmacology and physiology from forearm blood flow\cite{81}. Therefore, hand circulation is excluded by inflating a wrist cuff to suprasystolic pressure before measuring the forearm blood flow. For this reason, photoplethysmography is not used in measuring forearm VOP. However, the intention of forearm venous occlusion in this research is not to measure forearm blood flow, but is aimed at measuring changes in PPG DC signal, predominantly in the skin, as a consequence of changes in venous volume.

Each subject was given a 2-minute relaxation period before a disposable probe (Nellcor OxiMax Max-A/Max-AL, Nellcor, CA, USA) was attached on the left index finger. An occlusion cuff connected to a mercury sphygmomanometer was wrapped around the upper left arm, just above the elbow, before placing the same hand onto the non-contact PPG platform. After the subject felt comfortable with the setup of the devices, the experiment commenced with the DISCO4 acquiring resting data.

![Figure 4-30 Protocol Conducted Simultaneously Using Contact and Non-Contact PPG](image)

The PPG signal was first recorded for 60s, then the occlusion cuff was inflated to 40mmHg using a mercury sphygmomanometer and sustained for approximately 120s\cite{711}. The occlusion pressure of 40mmHg causes venous pooling in the arm. The low pressure exerted by the cuff would be sufficient to occlude the veins completely
but insufficient to obstruct the arterial inflow. Rapid deflation of the occlusion cuff would then restore the normal flow of venous blood. The timeline for this protocol is outlined in Figure 4-30.

The recorded data, which is the raw DC signal acquired from the DISCO4 platform, was processed using the moving average filter algorithm. Besides being an easy digital filter to implement, this filter is optimal for reducing random noise, thus making it favourable for time domain signals. The filtered output was then used to formulate the normalised VOP signal by normalising the entire DC sequence with the initial starting level of the DC signal.

Three VOP parameters were calculated using a PPG volume curve: venous capacitance, arterial inflow, and venous outflow parameters respectively. The increase in blood volume reduces the detected light transmission as a result of optical absorption, hence, an inversion of measured light intensity would reflect corresponding changes in blood volume. The purpose of assessing these parameters is to allow further comparison between contact and non-contact PPG systems. These data are used only as a comparison tool and are not measurable against established indices in the forearm as the adopted protocol was different. Although the protocol in this experiment is adapted from the VOP protocol used in the lower limb, the indices will be different as different limbs were studied.

The underlying principle of venous occlusion plethysmography is that arterial inflow is unaltered when a pressure exceeding venous pressure, but lower than arterial diastolic pressure, briefly interrupts the venous drainage. This results in the limb to swell at the rate equal to arterial inflow (AF), which can be determined from the rising slope in the VOP volume curve. The arterial inflow is calculated based on the tangent of the occlusion curve in the first three seconds after the start of the occlusion.

Owing to the blockade in venous return, the pressure distal to the cuff increases and will eventually rise to reach or exceed the occlusion pressure. At this pressure, the venous volume no longer increases, but will reach a plateau, which represents the capacity of venous system to store blood. This plateau, also known as venous
capacitance, is defined as the maximal venous distension at a specific pressure and is the measure of maximum change in limb blood volume from the baseline\textsuperscript{128}.

When the occlusion cuff is deflated rapidly, the outflow kinetics can be observed. Owing to the abrupt change in pressure, the congested blood will flow immediately out of the extremity, and thus a high rate of change will be obtained. The venous outflow is calculated as the tangent of the emptying curve in the first second after the release of occlusion cuff.

Venous outflow is the main parameter used in the diagnosis of deep vein thrombosis. A significant delay in the flow of blood upon deflation could indicate that the venous return is compromised. No delay was expected in this experiment, as all the chosen subjects were healthy, with no known vascular diseases.

4.6.1 Non-Contact Transmission Mode

An experiment was conducted to measure venous occlusion simultaneously using contact and non-contact transmission PPG probes on five healthy female subjects between the ages of 23 to 35 (28.5 ± 5.2 years), with no known vascular diseases. The subjects were asked to grip the fingers closely during the duration of the recording. Light intensity and gain were adjusted to produce a similar voltage level for both AC and DC signals for the two probes. The experiment was repeated 3 times on each subject.

Figure 4-31 shows the changes in light intensity measured during venous occlusion in one subject during the experiment. It can be observed that the two PPG systems have very similar responses to changes in blood volume owing to venous occlusion. The mean of the measured intensities, across each occlusion, for both PPG probes are tabulated in Table 4-6.
EXPERIMENTAL INVESTIGATION

Venous Occlusion Curve

Figure 4-31 VOP Curve Measured Simultaneously by Contact and Non-Contact Transmission PPG

<table>
<thead>
<tr>
<th>Data</th>
<th>Non-Contact PPG</th>
<th>Contact PPG</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Mean ± SD</td>
<td>Mean ± SD</td>
</tr>
<tr>
<td>1</td>
<td>0.985 ± 0.048</td>
<td>0.958 ± 0.078</td>
</tr>
<tr>
<td>2</td>
<td>0.946 ± 0.055</td>
<td>0.938 ± 0.072</td>
</tr>
<tr>
<td>3</td>
<td>0.924 ± 0.075</td>
<td>0.917 ± 0.075</td>
</tr>
<tr>
<td>4</td>
<td>0.905 ± 0.084</td>
<td>0.912 ± 0.073</td>
</tr>
<tr>
<td>5</td>
<td>0.960 ± 0.070</td>
<td>0.972 ± 0.068</td>
</tr>
<tr>
<td>6</td>
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<td>0.947 ± 0.071</td>
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<tr>
<td>7</td>
<td>0.961 ± 0.051</td>
<td>0.908 ± 0.075</td>
</tr>
<tr>
<td>8</td>
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<td>0.907 ± 0.092</td>
</tr>
<tr>
<td>9</td>
<td>0.916 ± 0.082</td>
<td>0.910 ± 0.091</td>
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<tr>
<td>10</td>
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<td>0.971 ± 0.089</td>
</tr>
<tr>
<td>11</td>
<td>0.929 ± 0.078</td>
<td>0.932 ± 0.084</td>
</tr>
<tr>
<td>12</td>
<td>0.972 ± 0.095</td>
<td>0.951 ± 0.101</td>
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<td>13</td>
<td>0.954 ± 0.064</td>
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</tr>
<tr>
<td>15</td>
<td>0.988 ± 0.064</td>
<td>0.934 ± 0.077</td>
</tr>
</tbody>
</table>

Table 4-6 Measure of Relative Mean DC of Contact and Non-Contact Transmission PPG
Three outliers have been removed because they have a much higher mean DC signal measured by the non-contact transmission PPG probe, which could be an indication that a strong presence of direct-coupled light was detected. These points were removed in order to portray a more precise analysis between signals measured by the two PPG probes.

Figure 4-32 shows a comparison of the mean relative amplitude of DC signal as measured by contact finger probe and by non-contact transmission PPG probe. The mean DC is 0.0033 with 95% CI of the mean at -0.0008 and 0.0074. The two limits of agreement at -0.0234 (95% CI at -0.0305 and -0.0163) and 0.0300 (95% CI at 0.0229 and 0.0371) are small enough to deduce that mean DC signal measured by the two systems are in agreement with one another.

Figure 4-33 shows a comparison of arterial inflow measured by a conventional contact PPG probe and the proposed non-contact PPG probe. The mean difference between the two systems in measuring arterial inflow is 0.23%/min with 95% limit agreement of the mean difference at -0.83%/min and 1.29%/min. Approximately two-third of the data sets are positioned close to the mean difference. When the arterial
inflow measured by non-contact transmission probe exhibits a lower reading, it indicates the presence of direct coupling as the arterial inflow can dictate the rate of change of the venous curve.

Figure 4-33  Arterial Inflow as Measured by Contact and Non-Contact Transmission PPG

Figure 4-34 displays the comparison of venous capacitance measured using the two PPG probes. The mean difference between the two PPG systems is 2.43%, which is low enough to argue that the difference between the two systems is small. Apart from one data set, higher venous capacitance readings were recorded using the contact PPG probe.

The mean difference of venous outflow, as shown in Figure 4-35, measured by the contact and non-contact PPG system is 9.26%/min with limits of agreement at 0.32%/min and 18.2%/min. Approximately 60% of the data sets are positioned close to the mean difference. The lack of agreement in the venous outflow could be caused by the increased detection of direct coupling and movement artefact. Movements of the skin might result in the occurrence of inter-digit gaps and hence, a detection of direct-coupled light. In the presence of direct coupling, the proportion of light intensity resulting from tissue decreases with an increasing detection of direct-coupled light.
EXPERIMENTAL INVESTIGATION

Figure 4-34  Venous Capacitance Measured by Contact and Non-Contact Transmission PPG

Figure 4-35  Venous Outflow as Measured by Contact and Non-Contact Transmission PPG
4.6.2 Non-Contact Reflection Mode

The experiment was conducted on fourteen subjects, seven male and seven female, between the age of 25 to 53 (31.50 ± 8.94) with no known vascular diseases to simultaneously measure venous occlusion plethysmography using both contact and non-contact reflection mode probe. A disposable probe (Nellcor OxiMax Max-A/Max-AL, Nellcor, CA, USA) was attached to left index finger and a mercury sphygmomanometer cuff was wrapped around the upper left arm. The left hand was then placed palm up on the PPG platform under the non-contact reflection probe.

The subjects were asked to relax but remain motionless during the duration of the recording to minimise the effect of movement artefact. The light intensity and gain for both the contact and non-contact probes were adjusted to generate similar voltage level for both the DC and AC components. The experiment followed closely the protocol that gave rise to results described in Figure 4-30, except that the PPG signal was only recorded for 30 seconds post deflation of the sphygmomanometer cuff.

![Venous Occlusion Curve](image)

*Figure 4-36 VOP Curve Measured Simultaneously Using Contact and Non-Contact Reflection PPG*
Figure 4-36 illustrates the simultaneous VOP curve as measured by contact and non-contact reflection PPG. The curves from the two systems were relatively different in amplitude and shape response. Table 4-7 records the relative mean (± standard deviation) for the DC signals measured for each of the subjects. One of the data sets has been discounted as the measured DC component for both contact and non-contact PPG was noisy and the VOP curve was not observed. This could be due to improper inflation of the cuff during measurement, that is, the cuff was not inflated rapidly enough.

<table>
<thead>
<tr>
<th>Data</th>
<th>Non-Contact PPG</th>
<th>Contact PPG</th>
</tr>
</thead>
<tbody>
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<td>Mean ± SD</td>
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<td>0.948 ± 0.057</td>
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<tr>
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</tr>
<tr>
<td>4</td>
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<td>0.866 ± 0.112</td>
</tr>
<tr>
<td>5</td>
<td>0.974 ± 0.020</td>
<td>0.900 ± 0.096</td>
</tr>
<tr>
<td>6</td>
<td>0.971 ± 0.027</td>
<td>0.907 ± 0.099</td>
</tr>
<tr>
<td>7</td>
<td>0.980 ± 0.018</td>
<td>0.926 ± 0.076</td>
</tr>
<tr>
<td>8</td>
<td>0.965 ± 0.026</td>
<td>0.901 ± 0.082</td>
</tr>
<tr>
<td>9</td>
<td>0.964 ± 0.025</td>
<td>0.912 ± 0.079</td>
</tr>
<tr>
<td>10</td>
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<td>0.902 ± 0.073</td>
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<tr>
<td>11</td>
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<td>0.897 ± 0.083</td>
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<tr>
<td>13</td>
<td>0.990 ± 0.028</td>
<td>0.939 ± 0.042</td>
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</table>

Table 4-7 Measure of Relative Mean DC of Contact and Non-Contact Reflection PPG

Figure 4-37 shows the Bland Altman plot to compare the VOP curve as measured by contact and non-contact reflection probe. The mean DC is 0.0612 (95% CI at 0.0555 and 0.0669), the lower limit agreement is 0.0206 (95% CI at 0.0196 and 0.0216) and the upper limit agreement is 0.1017 (95% CI at 0.1007 and 0.1027). All the data sets are located within the limit agreement, with approximately 70% of them located near the mean difference.
Figure 4-37 Mean DC Signal as Measured by Contact and Non-Contact Reflection PPG

Figure 4-38 Arterial Inflow as Measured by Contact and Non-Contact Reflection PPG

Figure 4-38 shows a comparison of arterial inflow calculated from the VOP curves that were measured by contact and non-contact reflection PPG. The mean difference between the arterial inflow measurements is -0.010%/min, the lower limit agreement is -0.032%/min and the upper limit of agreement is 0.012%/min. All but one of the
data points reside outside the limit of agreement, and it can be observed that the contact DC signal from these data displayed a comparatively much steeper curve than those from the non-contact reflection probe.

Figure 4-39  Venous Capacitance Measured by Contact and Non-Contact Reflection PPG

Figure 4-39 shows the comparison between venous capacitance calculated using the data sets of both contact and non-contact reflection PPG. The mean difference between the two PPG systems is 0.19%, while the lower limit of agreement is -0.04% and the upper limit of agreement is 0.41%. Whilst one of the data sets is outside the limit of agreement, 85% of the data sets are very close to the mean value. The data sets that are outside the limit shows a much higher measurement from non-contact reflection PPG signal that could be a result of reduced backscattered light due to reduced perfusion in the fingers.

Figure 4-40 displays the venous outflow as measured between contact and non-contact reflection probe. The mean difference of the venous outflow is -0.37%/min with limit agreement at -1.19%/min and 0.45%/min. Most of the data sets are scattered near the mean difference, while one of the data sets is out of the limit of agreement owing to some movement artefact that was detected during deflation.
In both of the performed experiments, venous occlusion in the forearm has generated a change to venous volume in the DC signal, and both the transmission and reflection non-contact probes have successfully detected the VOP curves. Although the light intensity in the two experiments was adjusted to a similar voltage level for both contact and non-contact probes, the degree of response between the non-contact transmission and non-contact reflection is different.

The VOP curve measured by non-contact transmission PPG probe is very similar to that measured by the contact probe, indicating that the non-contact transmission probe has a very similar response and is able to replicate signals that are measured by standard contact probes as both probes detect changes in PPG signal by measuring light absorption of the compounds in the tissue. However, the response between the contact probe and the non-contact reflection probe is relatively different. This could be explained by the fact that although the fingers have rich supply of blood vessels, it has relatively thick skin and it lacks skeletal structure to help reflect the illumination,
as reflection mode probes perform better on relatively thin skin that is close to skeletal structure with high density blood vessels.

Measurements of arterial inflow are influenced by the rate of inflation, and in non-contact transmission PPG, they are to some degrees influenced by the presence of direct coupling. The rate of cuff inflation, using a mercury sphygmomanometer, could influence the level of venous capacitance. A rapid change in pressure, during fast inflation of the cuff, can increase the velocity of blood flow towards the peripheries, hence, a higher venous pooling may occur. Nevertheless, both contact and non-contact PPG occlusion curves would have reflected the same responses to different effects of venous pooling because they were recorded simultaneously at the same site of measurement.

Venous capacitance indicates the maximum venous distension at a given pressure, and is the maximum volume recorded before deflation. Direct coupling influences changes in venous capacitance, as it is an amplitude-based measurement. When direct coupling is minimal, the rate of change in the non-contact DC signal increases, and this signal can have the same or even a higher measure of venous capacitance compared to contact PPG. An increase in direct coupling, on the other hand, reduces the percentage of measured intensity resulting from tissue, and as a result, a lower measure of venous capacitance. Owing to this, comparison of venous capacitance using a non-contact transmission probe reflected much higher values than those using a non-contact reflection probe.

The sudden removal of cuff pressure in the forearm results in a rapid decrease of venous pooling. This will instigate movements artefacts as the skin and tissue will attempt to swiftly restore its initial state. Hence, venous volume measured in the first second after deflation, by non-contact probe, could be compromised by the presence of direct coupling. This explains the higher measure of venous outflow obtained by the non-contact transmission probe. On the other hand, venous outflow measured using the non-contact reflection probe recorded a much lower mean difference and limit of agreement, as it is not influenced by direct-coupled light.
In this experiment, venous occlusion successfully generated a change to venous volume that enabled a comparison of DC component to be obtained between the contact and both the non-contact transmission and reflection probes. Direct coupling detected due to anatomical structure of the fingers can influence the signal intensity measured in the non-contact system and this is also reflected in measured VOP parameters. However, owing to the small presence of direct coupling, its effects on PPG signals are limited.

4.7 Example Applications Using Non-Contact Transmission PPG

The principles of operations for non-contact and conventional PPG are the same; the difference is in the type of probe used for acquiring measurements. Therefore, non-contact PPG can be used to replace a contact probe in principally all PPG applications. Two of the potential PPG applications are explored. As the thesis focuses on the investigation of the feasibility of non-contact PPG, in-depth analysis of the two PPG applications is not performed as part of the experiments but will be explored in detail in the future.

4.7.1 Peripheral Arterial Disease Assessment using Non-Contact PPG

Peripheral vascular disease is caused by structural changes, often narrowing or clogging, of the blood vessels outside the heart and brain. The most common subset is peripheral arterial disease (PAD). The most common assessment of PAD is the ankle-brachial pressure index (ABPI), which measures the systolic pressure ratio in the ankle and in the arm, and this index is regarded as a reliable predictor of lower extremity PAD.
Patients are usually requested to perform a pre-determined exercise for a short duration of time to estimate the extent of the disease. A significant drop of the pre- and post-exercise index could indicate a disorder arising from the arteries and further tests are then required. ABPI measured by conventional PPG is usually performed on the toes. However, when the toe is infected with ulcers, gangrene or has been amputated, PPG measurements can no longer be taken because of the restriction of the probe.

Nevertheless, if measurements were performed using non-contact PPG, this problem would not arise as the probe can be easily adjusted to measure perfusion from other parts of the foot instead. This system is more flexible, as it does not preclude patients just because of probe limitations. No ABPI measurements were performed at this stage. However, in a separate trial performed to analyse ABPI using a dual-channel PPG system, permission was given to record five sets of data using non-contact PPG on the foot.

During the measurements, care was taken to avoid the detection of direct coupling. Figure 4-41 shows two measurements made by non-contact transmission PPG on the ball of the foot during ABPI measurements. In the patient with a normal ABPI index, PPG signals were easily detected. However, in the patient with foreseen claudication (ABPI of 0.6), the signal is noisy, but some periodicity in the signals can be observed.

For this study, it is not possible to compare contact and non-contact PPG measurement owing to the use of a different PPG platform for recording. This platform only supports the use of a single probe for measurement and hence, does not allow for simultaneous recording of contact and non-contact probes. At this early stage, it is difficult to gauge the accuracy of the non-contact transmission PPG system in measuring ABPI, but a distinction in signals can be detected from healthy and possible claudication patients. In addition, the fact that non-contact PPG measurements can be performed on tissues other than the fingers has been shown.
4.7.2 Monitoring Respiration Rates by Non-Contact PPG

Cardiac variations, such as heart rate, stroke volume, arterial and venous pressure show fluctuations that are synchronous with respiration\(^{[129]}\). Transthoracic impedance pletysmography (TTI) is the ‘gold standard’ in clinical use today to measure respiratory rates. During both spontaneous and mechanical ventilation, respiration causes variations in peripheral circulation that can be influenced by a number of physiological factors.

A 24 year old female subject with no known vascular disease was asked to sit upright and relaxed on a chair. A disposable adhesive pulse oximeter sensor (Nellcor OxiMap Max-A/Max-AL, Nellcor, CA, USA) was carefully attached on the subject’s left index finger and the same hand was then positioned onto the non-contact PPG platform fixed with the non-contact transmission probe, ensuring that the fingers were closely gripped together with no apparent inter-digit gaps.
PPG signals were recorded simultaneously via contact and non-contact transmission PPG to observe the respiratory fluctuations in the signals. A 2-minute spontaneous breathing and 2-minute paced breathing at 6 cycles/min were recorded. Samples of 30s of the PPG signals are shown in Figure 4-42, with the timing for the pulse interval was shown in Figure 4-43. The frequency and amplitude variations in the PPG signals appeared to be similar to those recorded from spontaneous breathing. In paced breathing, the amplitude of PPG signals can be seen to oscillate in line with the respiration rate. Non-contact PPG is again seen to be noisier during diastole owing to increased tissue movements.

In order to verify the evaluation of respiratory fluctuations measured in non-contact PPG, comparisons have to be made with more established methods, such as TTI. The use of this probe is especially useful in neonatal care because it eliminates the use of probes on the sensitive skin of the neonates.

![Figure 4-42 Fluctuations of PPG Signal due to Spontaneous and Paced Breathing](image)
4.8 Summary

Various experiments were conducted in this research, each designed to determine the feasibility of the new method of non-contact photoplethysmography. In all the experiments, a custom designed PPG probe was used, together with the DISCO4 platform, to acquire non-contact PPG signals. Digitised PPG signals from LabJack were displayed to the user on the PC through a graphical user interface program written in LabView. Stored data were then processed offline using the Matlab software.

The first experiment was divided into three sub-protocols; measuring PPG signals using non-contact transmission probe on gripped fingers, on fingers with inter-digit gaps directly exposing the photodetector to the light source, and on fingers with inter-digits gaps without exposing the photodetector to the light source to investigate how light source positioning and tissue geometry can affect PPG signals.
The first protocol was compared with a prediction using a heuristic arterial model, which adequately predicts the normalised DC signal. The normalised AC signal in the prediction model showed a bigger change in signal amplitude with changes in light source positioning, believed to be related to contours of the fingers. Normalised AC signal in the measured data does not reflect this. It is believed that this is due to the limitations of the Beer-Lambert law, as it does not consider diffusion and scattering that occurs in the tissue.

An increase in probe distance and a decrease in the illumination strength, as the light source was positioned away from the photodetector, resulted in a gradual decrease of DC amplitude. As predicted, an increase in direct coupling resulted in a decrease in AC signal amplitude, the reason being that illumination resulting from direct coupling is stronger than illumination from tissue. However, it is shown that in small quantities direct coupling does not affect the quality of the AC signal.

In the strong presence of direct-coupled light in the second protocol, the measured AC signals were highly corrupted by motion artefacts and there might only be some very small integration of the AC signals that were detected. Nevertheless, when the light source was positioned away from direct illumination of the PIN photodiode, strong PPG signals were again detected; thus, positioning the light source away from direct-coupled light could avoid the detection of signals that are corrupted by motion artefacts.

In the third protocol, it was shown that the presence of direct coupling from any direction could affect the detection of AC signals. The quality of AC signals were uncertain and depended upon the proportion of direct coupling as well as the proportion resulting from the tissue during measurements. This experiment showed that direct coupling has to be avoided at all costs in order to consistently detect PPG signals that have satisfactory signal-to-noise ratio.

The second experiment was performed to explore how different geometry of the fingers can affect the measurement of non-contact reflection PPG signals. The experiment was divided into two sub-protocols; firstly, PPG signal was measured on
gripped fingers and secondly, PPG signal was measured on fingers with inter-digit gaps.

Good quality PPG signals were obtained using both protocols at all the offset positions. This shows that the presence of inter-digit gaps does not affect the quality of the PPG signal as direct coupling in reflection mode only occurs with inappropriate probe design.

The third and fourth experiments were conducted to demonstrate the feasibility of both the transmission and reflection non-contact PPG. This was accomplished by comparing the measurements made by each of the non-contact PPG probes with the conventional contact transmission-mode probe through simultaneous recordings.

The third experiment was performed in order to examine the similarity in AC signals measured between the contact and non-contact PPG probes. The systole of the contact and non-contact PPG signals correlate very strongly, but non-contact PPG signals are noisier during diastole. This could be a consequence of the expansion and constriction of the arteries during the heart cycle, thus resulting in random skin tissue movements and possibilities of small inter-digit gaps. However, this experiment showed that when there was reduced ambiguity in direct coupling, i.e. at the peaks during systole, non-contact PPG could provide measurements at a high degree of agreement with the conventional method.

In the fourth and final experiment, venous occlusion plethysmography was measured simultaneously using contact and both transmission and reflection non-contact PPG probes. There was a good agreement between each of the signals. The difference between the mean DC amplitude measured independently by the non-contact transmission and non-contact reflection probes with the contact probe is very low.

Three parameters (venous capacitance, arterial inflow and arterial outflow) were calculated based on the occlusion curves measured by the two PPG probes. These calculations showed good agreement between contact and non-contact reflection PPG probes for each of the parameters. However, the calculations of these parameters agreed less when using non-contact transmission probe. It is speculated that this is due
to the presence of direct coupling in the fingers, as no matter how tightly the fingers were gripped, small inter-digit gaps could still exist.

Hence, in this first stage of the feasibility study of non-contact transmission and reflection PPG, it is successfully demonstrated that it is indeed feasible to perform non-contact measurements that have strong agreements with the conventional method.

Two example applications of non-contact photoplethysmography have been explored. In both examples, PPG signal recorded using non-contact transmission PPG probe is comparable with those measured from contact probe. Further investigations are needed to validate these signals. However, it is illustrated that the non-contact probe has the potential of replacing the contact probe in the future.
PHOTONICS AND HEALTH TECHNOLOGY RESEARCH GROUP
LOUGHBOROUGH UNIVERSITY

CHAPTER FIVE

5  CONCLUSION
5.1 Conclusion

Non-invasive monitoring and diagnoses using plethysmographic techniques have various restrictions involving probe configuration and positioning, which limit their usefulness in many applications. PPG is the only plethysmographic technique that does not base its measurements on physical changes in the limb. PPG measures blood volume changes by optical variations in light-tissue absorption and hence, the only practical method can provide non-contact probe measurements. The aim of this thesis was to perform a feasibility study on the use of the non-contact probe in PPG measurements, with the hope it could in the future overcome the probe restrictions encountered by the current plethysmographic techniques.

Non-contact PPG operates on the same principle as the conventional PPG method, the only difference is in the use of the optoelectronics and probe positioning. In the non-contact transmission probe, a transmitter that consists of multiple LEDs and a receiver with a large PIN photodiode were used to improve the illumination and detection of non-contact PPG signals. These optoelectronics were encased in a reflector probes and was not attached directly to the tissue, but placed some distance away from the skin. For the non-contact reflection probe, multiple VCSELs were used to increase the light illumination so that there is enough illumination power to sufficiently penetrate into the skin tissue after travelling the distance between the probe and the skin tissue.

The custom-made sensor that caters for the increased illumination distance, and a PPG platform with high sensitivity for sensing small variations in measured signals were integrated in the proposed non-contact PPG system. Engineering issues that were addressed in the implementation of this system included the dynamic range of PPG signals, ambient artefacts and direct coupling.

The DISCO4 platform is adopted to deal with issues relating to dynamic range and ambient artefacts, as it has the flexibility to support applications that require extreme dynamic range and it is especially suited for sensing small variations signal in the presence of ambient light. Direct coupling, on the other hand, cannot be avoided when
using non-contact transmission probe especially if measurements are conducted in the peripheries, as it is influenced by the anatomical structure of the body parts.

The heuristic model developed in this thesis to describe non-contact PPG signals is based on the modification of the general contact PPG model that separates measured intensity into components from different optical paths. In non-contact transmission PPG, direct coupling cannot be ignored and is modelled into the equation as a function of light source positioning. The model showed that both the arterial and venous components in the PPG signals are affected by the presence of direct coupling and were later justified through experimental results. The dynamic AC term was normalised with respect to the position where the light source and photodetector are aligned to enable comparisons across the various light source positioning. The first order derivation of the model is influenced by the ratio of DC components originating from direct coupling and non-pulsatile tissue.

Various experiments were conducted to investigate the effects of direct coupling in non-contact PPG measurements and to ascertain the feasibility of measuring PPG signals using non-contact transmission and reflection PPG probes. The same hardware and software were used throughout the experiments. All off-line analyses were computed using Matlab.

In these experiments, it was shown that a good signal-to-noise ratio of PPG signals was successfully detected even when small measures of direct coupling were present. Although direct coupling is undesirable in both arterial and venous signals, it is unavoidable when measurements are acquired from well-perfused peripheries, such as fingers and toes, because of the anatomical structure of the vasculature. The results from various experiments have clearly shown that the new method of non-contact PPG was indeed feasible and has the potential to be developed further to overcome probe limitations in current plethysmography applications.
5.1.1 Non-Contact Transmission PPG

In the investigation of the effects of direct coupling on PPG signals using the non-contact transmission probe, it was proven that light source positioning and tissue geometry have a significant effect on the measured PPG signal. When non-contact measurements were conducted on closely gripped fingers to minimise the detection of direct-coupled light, measured light intensity displayed maximum illumination during alignment of the optoelectronics and a gradual decrease as the light source moved away from alignment. Direct-coupled light was detected on the left axis, but as the proportion of light resulting from direct coupling was small, the effects on the measured signals were also minimal. In all the predetermined light source positions, the recorded AC signals have good signal-to-noise ratios.

When non-contact PPG measurements were recorded on fingers with inter-digit gaps with the PIN photodiode fully exposed to the light source, the effects of direct coupling were evident when the light source was positioned near alignment, from 5mm offset on the left to 10mm offset to the right. No AC signals could be recorded in these positions, as the proportion of measured intensity resulting from direct coupling was too large. Instead, the direct-coupled light detected much of the skin tissue movements and hence, a high degree of motion artefacts were detected at these positions. However, when the light source was shifted away from the direct exposure, the reduced proportion of direct coupling enabled the PPG pulsatile signal to reappear.

When non-contact PPG measurements were recorded on fingers with inter-digit gaps, but on this occasion without directly exposing the photodetector to the light source, maximum light illumination was recorded at 5mm offset to the left owing to direct coupling. Motion artefacts were detected at some light source positions but these were unpredictable owing to the inconsistency in the inter-digit gaps positioning.

This experiment demonstrated that direct coupling could still occur in closely gripped fingers owing to the anatomical structure of the fingers themselves. Nevertheless, when the proportion of measured intensity resulting from direct coupling is low, acceptable signal-to-noise ratios of PPG signals were recorded. When the presence of
direct coupling is great, motion artefacts detected due to skin tissue movement would corrupt the small AC signals. Nevertheless, if the light source was positioned away from the direct coupling path, acceptable PPG signals could again be recorded.

Simultaneous recordings using contact and non-contact transmission probes performed to evaluate the feasibility of non-contact measurements showed promising results. The AC signal measured using the non-contact probe has a similar profile to that measured from the contact PPG probe, albeit being noisier during diastole because of skin surface movements. However, the amplitude positions of the PPG signals were unaffected, this was shown by the small mean differences between the contact and non-contact probes.

The small limit of agreement between contact and non-contact transmission AC PPG shows that signals recorded from the non-contact transmission probe is highly repeatable and has strong agreement with PPG signals recorded from the contact probe, and hence, can be used as a replacement for PPG applications using peak detection.

A change in DC is induced by using the venous occlusion plethysmography technique, to enable simultaneous comparisons of contact and non-contact PPG DC signals. Good agreement can be observed between the parameters measured between contact and non-contact transmission PPG probe.

5.1.2 Non-Contact Reflection PPG

In the experiment conducted to investigate the significance of tissue geometry on non-contact reflection PPG signal in two separate scenarios, it was observed that variation in the tissue geometry does not affect the quality of measured PPG signal. The PPG signal measured using a non-contact reflection probe exhibited good signal-to-noise ratio across all offset positions for measurements recorded from gripped fingers and fingers with inter-digit gaps.
Effects of direct coupling as seen in non-contact transmission measurement does not occur in non-contact reflection PPG signals because the light sources and the photodetector were shifted together as one unit as they are encompassed in the same probe, and hence, different tissue geometry positioning does not affect the measured PPG signal. However, optical shunting could occur within the non-contact reflection probe itself if it is not designed appropriately, and to avoid this, a VCSEL light source with narrow viewing angle is used.

The comparison of AC signals measured using contact and non-contact probes displayed very similar profiles. The non-contact reflection signal is less noisy compared to that recorded using non-contact transmission probe. The dichrotic notch can be clearly distinguished in the non-contact reflection signal. Owing to the strong agreement and repeatability, the non-contact reflection probe is a suitable substitution for a conventional contact probe.

The comparison of DC signals between contact and non-contact PPG recorded better agreement in measuring arterial inflow, venous capacitance and venous outflow parameters when using the non-contact reflection probe compared to the non-contact transmission probe. The reason for this finding is that non-contact reflection PPG is not affected by direct coupling. However, it is evident that there is a relatively significant difference in the venous occlusion DC profile measured by the contact and non-contact reflection probes. In conclusion, the overall degree of agreement obtained demonstrated that it is feasible to measure PPG signals using non-contact probes.

5.2 Future Work

This section outlines the suggestions on future work in this area, highlighting improvements that can be made to the current system and possible applications of the proposed method. The non-contact transmission and reflection PPG probes used throughout this research is the first prototype. The design was meant only as an
investigative tool to determine the feasibility of non-contact measurements and thus, there are several areas that can be improved upon.

5.2.1 New PPG Platform

The Photonics and Health Technology Research Group at Loughborough University has developed a new PPG platform known as the Venox board. This platform is a fully digitalised PPG driver and processor, has high sensitivity and is small in size. It is capable of supporting up to four different light source wavelengths and two different photodetectors.

The user has complete control over the setting of the light intensity and signal gain to suit the need of the user and the application. The Venox platform transmits and receives data using the standard RS232 protocol. It comes together with a graphical user interface that is written using Labview VI that enables real-time visualisation of the PPG signal and streaming of data.

The use of the new Venox board should improve the performance of the non-contact PPG probes because of the full-digitised functions. The compact size of the Venox board would potentially result in a PDA sized device that is easy to transport and use.

5.2.2 Improved Non-Contact Probe Design

In this feasibility study, all measurements were conducted on the tissue of the fingers, which are well-perfused because of the relatively thin surface dermal layer of the skin, thus enabling good light-tissue interactions.
5.2.2a Improving Current Non-Contact Transmission PPG System

Experimental results suggested that measurements from non-contact PPG, even with some degree of direct coupling, strongly agree with those obtained from the conventional method. Non-contact systems can however avoid the detection of direct coupling by measuring PPG signals at tissue areas that are not influenced by gaps, including the palm of the hand, the base of the foot and the wrist. The skin and tissue at these areas are considerably thicker in comparison to the tissues in the fingers, thus a light source with higher illumination is required. This can be achieved by designing a higher-powered light source that can supply larger measure of light intensity that is needed to penetrate through the thicker tissue area.

Larger reflectors can be utilised to maximise the light-tissue interactions in order to increase the spatial averaging that can reduce the effects of skin surface movements. Nevertheless, a larger reflector would also result in a larger proportion of ambient light and motion artefact detections, hence a better solution is required. Besides increasing the power of illumination and maximising the amount of light detection, the issue of motion artefacts from skin tissue movements need to be resolved. The possible use of another light source wavelength and use of an artefact reduction methodology is described in section 5.2.3.

The sensing range of the optoelectronics needs to be broadened to improve further the measurements made by non-contact PPG. Currently, the optimum separation of the probe is approximately 80mm. An increased sensing range is essential in non-contact PPG to eliminate the presence of the probe during measurements. Hence, optimal design and evaluation for an improved probe is required to develop the system further.

5.2.2b Improving Current Non-Contact Reflection PPG System

The feasibility study demonstrates that measurements from non-contact reflection PPG is not affected by direct coupling that is due to tissue geometry because a high quality PPG signal can be recorded even in the presence of inter-digit gaps. The
current sensing range for non-contact reflection PPG is 50mm. A further increase in sensing range is essential to improve the practical applicability of the non-contact reflection probe.

The current study is performed on the fingers that are rich with blood supply. Practically, non-contact reflection probe should be applicable on virtually any part of the body, and this includes some tissue areas that could have relatively thick skin where the arteries are located well below the skin, or the tissue area might not be well perfused, e.g. on patients with burns. In order to overcome these problems, a new probe design with higher intensity and output power would be needed to improve the system further. The increase in intensity and power would increase the sensing distance and would enable the illumination to penetrate deeper into the skin tissue. The design of the optical configuration can be improved to increase the collection of reflected light from the skin surface. Hence, optimisation of the design and evaluation for an improved probe is required to develop the system further.

5.2.3 Reducing the Effects of Skin Surface Movements

In the proposed method of non-contact photoplethysmography monitoring, the light source and photodetector are positioned away from the skin tissue to avoid any restrictions of probe placement and contact force\(^{[41,78]}\). The separation of optoelectronics from the skin tissue increases the likelihood of motion artefact detection from skin surface movements as well as the detection of direct coupling.

The presence of direct coupling in non-contact measurements could be caused by the expansion and constriction of the arteries, which in turn causes the skin surface to fluctuate. Involuntary movements, for example twitching and shivering, could also affect the profiles of the measured signals. These involuntary activities may be small, but because of the sensitivity of the non-contact probe, the illuminating light is able to detect these movements.
An artefact equalisation\textsuperscript{[130]} methodology can be developed to reduce the effects of intrinsic skin surface activities, by utilising an additional source wavelength that can provide sufficient information regarding these movements. It is suggested that investigation into this equalisation technique can be carried out using blue light, which operates within the wavelength range of 400-475\textnormal{nm}.

The specific extinction coefficient in this region is very high. When blue light is illuminated onto the tissue, the incident illumination will be very highly absorbed by blood. It will be very difficult for light to pass beyond the dermis because of the rich supply of blood vessels in this layer of tissue. A study performed using laser Doppler flowmetry has concluded that blue light (458\textnormal{nm}) investigates a more superficial blood flow, both \textit{in vitro} and \textit{in vivo}, when compared to a red wavelength (633\textnormal{nm})\textsuperscript{[131]}. Hence, if this speculation has merit, blue light should be able to detect motion artefacts due to skin surface movements and respiratory induced movements. Measured PPG signals are very noisy, as it is very sensitive to any tissue movements on the skin surface. An artefact equalisation model could then be developed to incorporate the use of blue light into the non-contact PPG probe to reduce the detection of movement artefacts.

5.2.4 Applications

The principles of operation for non-contact and conventional PPG are the same; the only difference will be in the type of probe used for acquiring measurements. Therefore, non-contact PPG can be used in principally all conventional PPG applications. A few of these applications are listed in the following sub-topics.

5.2.4a \textit{Venous Occlusion Plethysmography using Non-Contact PPG}

This research has conducted various measurements of venous occlusion plethysmography using non-contact PPG. However, measurements were made in the
forearm instead of the lower limb. The protocol for VOP measurements in the limb using non-contact PPG will be similar to the protocol applied in this research.

In order to improve the measure of VOP, an automatic pressure measurement device can be used. An automatic device can ensure a constant duration and a steady applied pressure during inflation and occlusion of the cuff across all measurements. The E20 Rapid Cuff Inflator (by Hokanson, Washington, US) may be a suitable device for this purpose. It provides rapid inflation and has an optional timer that allows the programmable rate of inflation and deflation of the cuff. In addition, the E20 also displays the cuff pressure within 1mmHg.

It is essential to obtain measurements that are comparable against standardised indices. The presence of direct coupling can be eliminated by calibrating the venous occlusion curve with measurements from another light source (Section 5.2.3). On the other hand, measurements using non-contact PPG could be conducted on the foot away from the toes to avoid direct coupling completely. No measurements were performed at this time owing to the lack of an automated pressure inflator.

5.2.4b Feature Extraction of Non-Contact PPG Pulse

The peripheral pulse is often used in the assessment of health and disease because it can provide information regarding the cardiovascular system and hence properties of blood vessels. PPG is one of the various techniques that can be used to measure peripheral pulses, which can provide valuable information about peripheral circulation.

Studies have shown that PPG pulses can be used to detect peripheral vascular disease but there is more ambiguity in describing how age and vascular properties affects the peripheral pulse at various body sites. These studies include quantifying pulse transit timing, variations of pulse wave velocity with age and arterial stiffness, as well as demonstrating age-related changes in shape characteristics.
Pulse shapes have common features, a systolic rising edge and a dichrotic notch attributed to the closure of the aortic valve and wave reflection from the periphery. It has shown that PPG pulses may become smoothed and elongated because of changes in resistance and compliance of the arteries with advancing age. The fact that the dichrotic notch diminishes in older subjects can be partly attributed to the increased pulse wave velocity.

5.2.5 Future Clinical Applications

In this section, possible new applications of non-contact PPG are described. One of the applications involves measuring tissue perfusion in patients with burns using a non-contact PPG probe and the other is a smart bed equipped with non-contact optoelectronics.

5.2.5a Burns Monitoring

Burns can vary significantly in their severity and can be classed as superficial burns, partial-thickness burns and full-thickness burns. Superficial (1st degree) burns involve only the epidermis layer of the skin, while partial-thickness (2nd degree) burns involve damage to the epidermis, papillary and dermis layers of the skin. Full thickness (3rd degree) burns involve all layers of epidermis and dermis, and may even extend to the subcutaneous structures. Factors that can determine the burn depth are temperature, duration of contact, blood flow to the skin and anatomic location.

Burns can cause oedema to form quickly in the tissue owing to leakage of the capillaries. Oedema in the limb may cause ischaemia, which can lead to limb loss. In addition, burns can cause hypovolaemia, which leads to poor systemic tissue perfusion. This condition, known as burn shock, is the inability of the circulation to meet the needs of tissues for oxygen, nutrients and waste removal.
The estimation of burns is frequently based on the ‘rule of nines’, where the body is divided into nine areas, each with a different value of body surface area rating. This method can underestimate burn size in infants and children. Another approximation of burns can be made using a variation of the Lund and Browder burn size chart that takes into account the different proportions of the body at various ages. Both methods can underestimate the degree of burns because they depend on visual inspections made by the physician as well as the experience of the physician himself.

In order to obtain a more precise estimation, non-contact PPG can be used for perfusion measurements, as the probe will not be in contact with the skin. Non-contact PPG can identify the extent of burns in the tissues by the difference in the measured perfusion. This can help in prioritising wound care to areas that are badly damaged. Besides estimating the degree of burns, non-contact PPG can also be used to monitor the healing process by monitoring the perfusion to the tissue areas.

The implications of using this technique to monitor patients with burns are yet to be established. Further research would need to be carried out in this area in order to establish the viability of measuring tissue perfusion in burns victims using non-contact PPG.

5.2.5b Smart Bed

The idea of a smart bed is to have a bed fitted with optoelectronics that can enable PPG measurements using a non-contact system. The user will just need to switch on the monitoring system before using the bed. A monitor will be used to display the various states of measurements. It can also be equipped with an alarm system that will be triggered when measurements go below the preset threshold.

This bed can be very useful in the intensive care unit. It can be equipped for various types of monitoring including heart rate, respiration and tissue oxygenation. The use of a smart bed helps to reduce the different types of equipment that are needed to make these measurements, hence making ICU less hostile and intimidating. In
addition, a smart bed reduces the strain on nurses who have to ensure proper attachment of the equipment and it reduces the workload of technicians who have to service and calibrate this equipment.

Besides ICU, people with sleep apnoea can also use the smart bed to monitor respiration and heart rates. In people with sleep apnoea, upper airway obstruction causes temporary stoppage of breathing during sleep. The smart bed can then be used to monitor these occurrences so that the physician can be better informed of the condition. In addition, the smart bed can also raise an alarm when the condition worsens.

The smart bed could also be very useful in neonatal and paediatric care. It is often hard to attach equipment on neonates and young children. Young children do not like to be attached to equipment because of the restriction to their movement, so they will often try to detach the equipment. With the smart bed, monitoring can be easily carried out because there will be no equipment attached directly to them. In addition, it can be hard to attach equipment to neonates because of their fragile nature and they too will try to detach any equipment when it is found to be uncomfortable. The smart bed can be used to solve this problem, as well as to reduce the worries of parents who have to endure the situation.

However, the smart bed is still an idea that can possibly emerge from non-contact monitoring. One of the major barriers in the implementation of this application is movement artefact. The most plausible solution is to secure several sets of sensors around the bed, with the intention that some of these sensors will still record quality measurement when movement occurs in part of the body. Much more research and development will be needed in this area to provide accurate measurements before this idea can be realised. The smart bed can indeed improve the primary health care service if it is proved to be feasible.
REFERENCES


REFERENCES


REFERENCES


REFERENCES


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Kingbright®

1.6x0.8mm SMD CHIP LED LAMPS
KP-1608 SERIES

Features
- 1.6mmx0.8mm SMT LED. 1.1mm THICKNESS.
- LOW POWER CONSUMPTION.
- WIDE VIEWING ANGLE.
- IDEAL FOR BACKLIGHT AND INDICATOR.
- VARIOUS COLORS AND LENS TYPES AVAILABLE.

Description
The Bright Red source color devices are made with Gallium Phosphide Red Light Emitting Diode.

The Super Bright Green source color devices are made with Gallium Phosphide Green Light Emitting Diode.

The High Efficiency Red source color devices are made with Gallium Arsenide Phosphide on Gallium Phosphide Orange Light Emitting Diode.

The Yellow source color devices are made with Gallium Arsenide Phosphide on Gallium Phosphide Yellow Light Emitting Diode.

The Super Bright Red source color devices are made with Gallium Aluminum Arsenide Red Light Emitting Diodes.

Selection Guide

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Notes:
1. All dimensions are in millimeters (inches).
2. Tolerance is ±0.1 (0.004) unless otherwise noted.
3. Lead spacing is measured where the lead emerge the package.
4. Specifications are subject to change without notice.

Note: 1. 1° is the angle from optical centerline where the luminous intensity is 1/2 the optical centerline value.

1-KP-1608
### Electrical / Optical Characteristics at $T_A=25^\circ C$

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<td>Super Bright Green</td>
<td>660</td>
<td>565</td>
<td></td>
<td></td>
</tr>
<tr>
<td>$\Delta\lambda/2$</td>
<td>Spectral Line Halfwidth</td>
<td>Bright Red</td>
<td>45</td>
<td>45</td>
<td>nm</td>
<td>IF=20mA</td>
</tr>
<tr>
<td></td>
<td></td>
<td>High Efficiency Red</td>
<td>45</td>
<td>45</td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td>Yellow</td>
<td>45</td>
<td>35</td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td>Super Bright Red</td>
<td>35</td>
<td>20</td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td>Super Bright Green</td>
<td>20</td>
<td>30</td>
<td></td>
<td></td>
</tr>
<tr>
<td>C</td>
<td>Capacitance</td>
<td>Bright Red</td>
<td>40</td>
<td>12</td>
<td>pF</td>
<td>VF=OV,f=1MHz</td>
</tr>
<tr>
<td></td>
<td></td>
<td>High Efficiency Red</td>
<td>40</td>
<td>12</td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td>Yellow</td>
<td>12</td>
<td>10</td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td>Super Bright Red</td>
<td>10</td>
<td>95</td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td>Super Bright Green</td>
<td>95</td>
<td>45</td>
<td></td>
<td></td>
</tr>
<tr>
<td>$V_F$</td>
<td>Forward Voltage</td>
<td>Bright Red</td>
<td>2.0</td>
<td>2.5</td>
<td>V</td>
<td>IF=20mA</td>
</tr>
<tr>
<td></td>
<td></td>
<td>High Efficiency Red</td>
<td>2.0</td>
<td>2.5</td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td>Yellow</td>
<td>2.1</td>
<td>2.5</td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td>Super Bright Red</td>
<td>2.1</td>
<td>2.5</td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td>Super Bright Green</td>
<td>2.2</td>
<td>2.5</td>
<td></td>
<td></td>
</tr>
<tr>
<td>$I_R$</td>
<td>Reverse Current</td>
<td>All</td>
<td>10</td>
<td></td>
<td>uA</td>
<td>VR = 5V</td>
</tr>
</tbody>
</table>

### Absolute Maximum Ratings at $T_A=25^\circ C$

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Bright Red</th>
<th>High Efficiency Red</th>
<th>Yellow</th>
<th>Super Bright Red</th>
<th>Super Bright Green</th>
<th>Units</th>
</tr>
</thead>
<tbody>
<tr>
<td>Power dissipation</td>
<td>105</td>
<td>105</td>
<td>105</td>
<td>100</td>
<td>105</td>
<td>mW</td>
</tr>
<tr>
<td>DC Forward Current</td>
<td>25</td>
<td>30</td>
<td>30</td>
<td>30</td>
<td>25</td>
<td>mA</td>
</tr>
<tr>
<td>Peak Forward Current [1]</td>
<td>150</td>
<td>30</td>
<td>150</td>
<td>30</td>
<td>25</td>
<td>mA</td>
</tr>
<tr>
<td>Reverse Voltage</td>
<td>5</td>
<td>5</td>
<td>5</td>
<td>5</td>
<td>5</td>
<td>V</td>
</tr>
<tr>
<td>Operating/Storage Temperature</td>
<td>-40°C To +85°C</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Lead Soldering Temperature [2]</td>
<td>230°C For 3 Seconds</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

Note:
1. 1:10 Duty Cycle, 0.1ms Pulse Width.
Yellow KP-1608YD, KP-1608YC, KP-1608YT

Super Bright Red KP-1608SRD, KP-1608SRC, KP-1608SRT
Super Bright Green KP-1608SGD, KP-1608SGC, KP-1608SGT

FORWARD CURRENT vs. FORWARD VOLTAGE

FORWARD CURRENT DERATING CURVE

LUMINOUS INTENSITY vs. FORWARD CURRENT

SPATIAL DISTRIBUTION

KP-1608 Series SMT Reflow Soldering Instructions

Temperature

Time

230°C

140-160°C

5°C/sec max.

-5°C/sec max.

5°C/sec max.

OVER 120sec. 125sec.(STANDARD)

5 sec max.
KP-160B Series Recommended Soldering Pattern

FOR REFLOW SOLDERING

KP-160B Series Tape Specifications

*SR type polarity is opposite

(Unit: mm)

(Units: mm)
APPENDIX B

Data Sheet for Surface Mount Infrared LED

Model KP-2012SF4C
### Selection Guide

<table>
<thead>
<tr>
<th>Part No.</th>
<th>Dice</th>
<th>Lens Type</th>
<th>$P_{o}$ (mW/sr) [2] @ 20mA</th>
<th>Viewing Angle [1]</th>
</tr>
</thead>
<tbody>
<tr>
<td>KP-2012SF4C</td>
<td>SF4 GaAlAs</td>
<td>WATER CLEAR</td>
<td>0.4</td>
<td>120°</td>
</tr>
</tbody>
</table>

Notes:
1. 91/2 is the angle from optical centerline where the luminous intensity is 1/2 the optical centerline value.
2. Radiant intensity/luminous flux: +/-15%.

### Electrical / Optical Characteristics at $TA=25^\circ C$

<table>
<thead>
<tr>
<th>Parameter</th>
<th>P/N</th>
<th>Symbol</th>
<th>Typ.</th>
<th>Max.</th>
<th>Units</th>
<th>Test Conditions</th>
</tr>
</thead>
<tbody>
<tr>
<td>Forward Voltage [1]</td>
<td>SF4</td>
<td>$V_F$</td>
<td>1.3</td>
<td>1.6</td>
<td>V</td>
<td>$I_F=20mA$</td>
</tr>
<tr>
<td>Reverse Current</td>
<td>SF4</td>
<td>$I_R$</td>
<td></td>
<td>10</td>
<td>$\mu$A</td>
<td>$V_R=5V$</td>
</tr>
<tr>
<td>Capacitance</td>
<td>SF4</td>
<td>$C$</td>
<td>80</td>
<td></td>
<td>pF</td>
<td>$V_R=0V; f=1MHz$</td>
</tr>
<tr>
<td>Peak Spectral Wavelength</td>
<td>SF4</td>
<td>$\lambda_P$</td>
<td>880</td>
<td></td>
<td>nm</td>
<td>$I_F=20mA$</td>
</tr>
<tr>
<td>Spectral Bandwidth</td>
<td>SF4</td>
<td>$\Delta \lambda/2$</td>
<td>50</td>
<td></td>
<td>nm</td>
<td>$I_F=20mA$</td>
</tr>
</tbody>
</table>

Note:
1. Forward Voltage: +/-0.1V.

### Absolute Maximum Ratings at $TA=25^\circ C$

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Symbol</th>
<th>SF4</th>
<th>Units</th>
</tr>
</thead>
<tbody>
<tr>
<td>Power dissipation</td>
<td>$P_T$</td>
<td>80</td>
<td>mW</td>
</tr>
<tr>
<td>DC Forward Current</td>
<td>$I_F$</td>
<td>50</td>
<td>mA</td>
</tr>
<tr>
<td>Peak Forward Current [1]</td>
<td>$I_{FS}$</td>
<td>1.2</td>
<td>A</td>
</tr>
<tr>
<td>Reverse Voltage</td>
<td>$V_R$</td>
<td>5</td>
<td>V</td>
</tr>
<tr>
<td>Operating Temperature</td>
<td>$T_A$</td>
<td>-40 To +85</td>
<td>°C</td>
</tr>
<tr>
<td>Storage Temperature</td>
<td>$T_{STG}$</td>
<td>-40 To +85</td>
<td>°C</td>
</tr>
</tbody>
</table>

Note:
1. 1/100 Duty Cycle, 10us Pulse Width.
NOTES:
1. We recommend the reflow temperature 245°C(±5°C) 2. maximum soldering temperature should be limited to 260°C 3. Don't cause stress to the epoxy resin while it is exposed to high temperature. 4. Shrinkage of the reflow process shall be 2 times or less.

Recommended Soldering Pattern
(Units: mm; Tolerance: ±0.1)

Tape Specifications
(Units: mm)
APPENDIX C

Data Sheet for Surface Mount Photodetector

Model OSD15-5T
Silicon Photodetector

BLUE SENSITIVE FOR BIASED OR UNBIASED OPERATION

The Centronic Series 5T detectors offer high blue sensitivity coupled with high shunt resistance and low dark leakage current. They are particularly suited to low light level applications from 430-900 nm where the highest signal to noise ratio is important. They may be operated photovoltaically or with a reverse bias of up to 12V where lower capacitance is needed. The 5T range provides the most economic solution for all applications where high speed of response above 800 nm is not critical.

ABSOLUTE MAXIMUM RATINGS

<table>
<thead>
<tr>
<th></th>
<th>Max. Rating</th>
</tr>
</thead>
<tbody>
<tr>
<td>DC Reverse Voltage</td>
<td>15V</td>
</tr>
<tr>
<td>Peak Pulse Current (1 μs, 1% duty cycle)</td>
<td>200mA</td>
</tr>
<tr>
<td>Peak DC Current</td>
<td>10mA</td>
</tr>
<tr>
<td>Storage Temperature Range</td>
<td></td>
</tr>
<tr>
<td>Except for those listed below:</td>
<td></td>
</tr>
<tr>
<td>-45°C to + 100°C</td>
<td></td>
</tr>
<tr>
<td>-25°C to + 80°C</td>
<td></td>
</tr>
<tr>
<td>Operating Temperature Range</td>
<td></td>
</tr>
<tr>
<td>Except for: LD12, LD16, LD20, LD35, MD25, MD100 and MD144-5T</td>
<td></td>
</tr>
<tr>
<td>-25°C to + 75°C</td>
<td></td>
</tr>
<tr>
<td>0°C to + 75°C</td>
<td></td>
</tr>
<tr>
<td>Soldering Temperature for 5 seconds max.</td>
<td>200°C</td>
</tr>
</tbody>
</table>

Series 5T – Typical Spectral Response

Series 5T – Typical Capacitance versus Bias Voltage for a given Detector Area

WWW: www.centronic.co.uk
email: esales@centronic.co.uk
Tel: +44(0)1689 808022
Fax: +44(0)1689 845117

APPENDICES
Electrical / Optical Specifications

Characteristics measured at 22°C (±2) ambient, and a reverse bias of 12 volts, unless otherwise stated. Shunt Resistance measured at ± 10 mV.

For rise time on Quadrants, Linear and Matrix Arrays take figures for single element diodes having equivalent active area.

### Single Elements

<table>
<thead>
<tr>
<th>Type No.</th>
<th>Active Area (Total)</th>
<th>Responsivity A/W</th>
<th>Dark Current (nA)</th>
<th>NEP WHz-%</th>
<th>Capacitance pF</th>
<th>Shunt Resistance Megohms</th>
<th>Rise time ns</th>
<th>Package</th>
</tr>
</thead>
<tbody>
<tr>
<td>OSD1-5T</td>
<td>1 x 1.13 dia</td>
<td>0.18</td>
<td>0.21</td>
<td>1</td>
<td>0.2</td>
<td>2.5 x 10^-14</td>
<td>35</td>
<td>7</td>
</tr>
<tr>
<td>OSD3-5T</td>
<td>3 x 2.18 x 1.4</td>
<td>0.18</td>
<td>0.21</td>
<td>2</td>
<td>0.5</td>
<td>3.0 x 10^-14</td>
<td>80</td>
<td>20</td>
</tr>
<tr>
<td>OSD5-5T</td>
<td>5 x 2.52 dia</td>
<td>0.18</td>
<td>0.21</td>
<td>2</td>
<td>0.5</td>
<td>3.3 x 10^-14</td>
<td>130</td>
<td>35</td>
</tr>
<tr>
<td>OSD7-5T</td>
<td>7.5 x 2.75 x 2.75</td>
<td>0.18</td>
<td>0.21</td>
<td>3</td>
<td>1</td>
<td>4.6 x 10^-14</td>
<td>180</td>
<td>40</td>
</tr>
<tr>
<td>OSD10-5T</td>
<td>15 x 3.8 x 3.8</td>
<td>0.18</td>
<td>0.21</td>
<td>5</td>
<td>1</td>
<td>5.5 x 10^-14</td>
<td>300</td>
<td>80</td>
</tr>
<tr>
<td>OSD35-5T</td>
<td>35 x 5.9 x 5.9</td>
<td>0.18</td>
<td>0.21</td>
<td>10</td>
<td>2</td>
<td>7.5 x 10^-14</td>
<td>950</td>
<td>200</td>
</tr>
<tr>
<td>OSD50-5T</td>
<td>50 x 7.98 dia</td>
<td>0.18</td>
<td>0.21</td>
<td>15</td>
<td>5</td>
<td>1.6 x 10^-13</td>
<td>1300</td>
<td>270</td>
</tr>
<tr>
<td>OSD60-5T</td>
<td>62 x 7.9 x 7.9</td>
<td>0.18</td>
<td>0.21</td>
<td>25</td>
<td>6</td>
<td>2.3 x 10^-13</td>
<td>1800</td>
<td>310</td>
</tr>
<tr>
<td>OSD100-5T</td>
<td>100 x 14.3 dia</td>
<td>0.18</td>
<td>0.21</td>
<td>60</td>
<td>8</td>
<td>2.1 x 10^-13</td>
<td>2500</td>
<td>520</td>
</tr>
<tr>
<td>OSD300-5T</td>
<td>300 x 19.54 dia</td>
<td>0.18</td>
<td>0.21</td>
<td>200</td>
<td>30</td>
<td>3.5 x 10^-13</td>
<td>7500</td>
<td>1500</td>
</tr>
</tbody>
</table>

### Quadrants

(Values given are per element unless otherwise stated)

<table>
<thead>
<tr>
<th>Type No.</th>
<th>Active Area (Total)</th>
<th>Responsivity A/W</th>
<th>Dark Current nA</th>
<th>NEP WHz-%</th>
<th>Capacitance pF</th>
<th>Shunt Resistance Megohms</th>
<th>Crosstalk %</th>
<th>Package</th>
</tr>
</thead>
<tbody>
<tr>
<td>QD7-5T</td>
<td>7 x 2.99 dia</td>
<td>0.18</td>
<td>0.21</td>
<td>6</td>
<td>2</td>
<td>2.3 x 10^-14</td>
<td>50</td>
<td>15</td>
</tr>
<tr>
<td>QD50-5T</td>
<td>50 x 7.98 dia</td>
<td>0.18</td>
<td>0.21</td>
<td>30</td>
<td>3</td>
<td>4.6 x 10^-14</td>
<td>330</td>
<td>80</td>
</tr>
<tr>
<td>QD100-5T</td>
<td>100 x 11.3 dia</td>
<td>0.18</td>
<td>0.18</td>
<td>50</td>
<td>5</td>
<td>7.0 x 10^-14</td>
<td>650</td>
<td>130</td>
</tr>
</tbody>
</table>
### Linear Arrays

(Values given are per element unless otherwise stated)

<table>
<thead>
<tr>
<th>Type No.</th>
<th>No. of Elements</th>
<th>Array Dimensions</th>
<th>Responsivity A/W $\lambda = 436$ nm $V_r = 0V$</th>
<th>Shunt Resistance Megohms</th>
<th>NEP WHz$^{-1/2}$ $\lambda = 436$ nm</th>
<th>Capacitance pF</th>
<th>Dark Current nA</th>
</tr>
</thead>
<tbody>
<tr>
<td>LD2A-5T</td>
<td>2</td>
<td>1.00 x 2.0 x 0.5</td>
<td>0.05</td>
<td>0.18</td>
<td>0.21</td>
<td>100</td>
<td>1000</td>
</tr>
<tr>
<td>LD2B-5T</td>
<td>2</td>
<td>2.02 x 1.42 x 0.45</td>
<td>0.18</td>
<td>0.21</td>
<td>50</td>
<td>1000</td>
<td>$2.5 \times 10^{-14}$</td>
</tr>
<tr>
<td>LD2C-5T</td>
<td>2</td>
<td>4.83 x 1.27 x 0.38</td>
<td>0.05</td>
<td>0.15</td>
<td>0.18</td>
<td>100</td>
<td>1000</td>
</tr>
<tr>
<td>LD4A-5T</td>
<td>4</td>
<td>15.0 x 3.0 x 5.0</td>
<td>0.05</td>
<td>0.15</td>
<td>0.18</td>
<td>15</td>
<td>400</td>
</tr>
<tr>
<td>LD4B-5T</td>
<td>4</td>
<td>0.64 x 0.8 x 0.3</td>
<td>0.15</td>
<td>0.18</td>
<td>40</td>
<td>500</td>
<td>$4.0 \times 10^{-14}$</td>
</tr>
<tr>
<td>LD5A-5T</td>
<td>5</td>
<td>0.10 x 0.125 x 0.8</td>
<td>0.05</td>
<td>0.15</td>
<td>0.18</td>
<td>100</td>
<td>1000</td>
</tr>
<tr>
<td>LD12A-5T</td>
<td>12</td>
<td>0.25 x 0.5 x 0.5</td>
<td>0.05</td>
<td>0.15</td>
<td>0.18</td>
<td>100</td>
<td>2000</td>
</tr>
<tr>
<td>LD16C-5T</td>
<td>16</td>
<td>0.035 x 0.2 x 0.175</td>
<td>0.025</td>
<td>0.15</td>
<td>0.18</td>
<td>100</td>
<td>2000</td>
</tr>
<tr>
<td>LD16(1.5)-ST 16</td>
<td>1.8 x 2.1</td>
<td>0.9 x 0.1</td>
<td>0.18</td>
<td>0.21</td>
<td>100</td>
<td>1500</td>
<td>$2.0 \times 10^{-14}$</td>
</tr>
<tr>
<td>LD16(2.5)-ST 16</td>
<td>2.5 x 2.5</td>
<td>1 x 0.5</td>
<td>0.18</td>
<td>0.21</td>
<td>100</td>
<td>1500</td>
<td>$2.0 \times 10^{-14}$</td>
</tr>
<tr>
<td>LD20-5T</td>
<td>20</td>
<td>3.50 x 4.0 x 0.9</td>
<td>0.05</td>
<td>0.15</td>
<td>0.18</td>
<td>100</td>
<td>1000</td>
</tr>
<tr>
<td>LD20(0.36)-ST 20</td>
<td>0.36 x 0.6 x 0.6</td>
<td>0.1</td>
<td>0.18</td>
<td>0.21</td>
<td>100</td>
<td>2000</td>
<td>$1.7 \times 10^{-14}$</td>
</tr>
<tr>
<td>LD35ST</td>
<td>35</td>
<td>4.42 x 4.6 x 0.9</td>
<td>0.03</td>
<td>0.18</td>
<td>0.21</td>
<td>40</td>
<td>2000</td>
</tr>
</tbody>
</table>

### Matrix Arrays

(Values given are per element unless otherwise stated)

<table>
<thead>
<tr>
<th>Type No.</th>
<th>No. of Elements</th>
<th>Array Dimensions</th>
<th>Responsivity A/W $\lambda = 438$ nm $V_r = 0V$</th>
<th>Shunt Resistance Megohms</th>
<th>NEP WHz$^{-1/2}$ $\lambda = 438$ nm</th>
<th>Capacitance pF</th>
<th>Dark Current nA</th>
</tr>
</thead>
<tbody>
<tr>
<td>MD25-5T</td>
<td>5 x 5</td>
<td>7.99 x 2.7 x 2.7</td>
<td>0.1</td>
<td>0.15</td>
<td>0.18</td>
<td>5</td>
<td>2000</td>
</tr>
<tr>
<td>MD100-5T</td>
<td>10 x 10</td>
<td>1.96 x 1.4 x 1.4</td>
<td>0.1</td>
<td>0.15</td>
<td>0.18</td>
<td>1</td>
<td>400</td>
</tr>
<tr>
<td>MD144-5T</td>
<td>12 x 12</td>
<td>1.96 x 1.4 x 1.4</td>
<td>0.1</td>
<td>0.15</td>
<td>0.18</td>
<td>1</td>
<td>400</td>
</tr>
</tbody>
</table>

**Note:** Recommended operating voltage range 0 to 12 volts, for all Series 5T Detectors

Highlighted items are Centronic standard products generally available from stock.
APPENDIX D

Data Sheet for Set Screw Reflector

Model 1938
Reflectors

Carley Lamps manufactures Aluminum Reflectors, Set Screw Reflector, and Lamp Sleeves/Reflector Sleeves designed to fit Carley's Set Screw Assemblies.

- Aluminum Reflectors
- Set Screw Reflectors
- Lamp Sleeves/Reflector Sleeves

Aluminum Reflectors

Carley Lamps manufactures parabolic and elliptical reflectors ranging from .25 inches to 3.00 inches in diameter. Carley Lamps can produce reflectors up to 6.0" in diameter, as well as offer several coatings and textures on the reflector face. Carley reflectors are hand polished, and vacuum metallized coated when required. Hand polishing reflectors forms a fine coating of aluminum oxide on the reflector face and this protects the reflector, maintaining its bright surface. Carley Aluminum reflectors are excellent at absorbing heat and they hold up against high wattage lamps better than plastic reflectors. Carley aluminum reflectors are excellent at reflecting, collimating and refocusing both visible, ultra violet and Infra Red Light. Contact Carley Lamps today for assistance in choosing the best performing reflector for your application.

<table>
<thead>
<tr>
<th>REFLECTOR</th>
<th>TYPE</th>
<th>DIA.</th>
<th>OIL</th>
<th>REAM</th>
<th>REFOCU</th>
<th>RFOCAL</th>
<th>FOR LAMP</th>
</tr>
</thead>
<tbody>
<tr>
<td>1578</td>
<td>Ell</td>
<td>.375</td>
<td>.50</td>
<td>128</td>
<td>18</td>
<td>.05</td>
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<tr>
<td>1579</td>
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<td>.375</td>
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<td>128</td>
<td>18</td>
<td>.05</td>
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<tr>
<td>1580</td>
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<td>.50</td>
<td>189</td>
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<td>.75</td>
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<td>1583</td>
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<td>1584</td>
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<td>.75</td>
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<td>2101</td>
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<td>343</td>
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<tr>
<td>2102</td>
<td>Ell</td>
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<td>2.50</td>
<td>343</td>
<td>1.20</td>
<td>T-2 1/2</td>
<td></td>
</tr>
</tbody>
</table>

* Par = Parabolic (Collimating) Ell = Elliptical (Refocusing).

** Optimum Refocus - refocus can be changed by moving lamp forward or backward inside the reflector.

NOTE: By locating the lamp filament at the "focal point," you will obtain maximum performance. Securing the lamp to a standard reflector is normally accomplished by gluing the lamp into the reflector. For greater options and ease of use, review the Set Screw Reflectors.

Parabolic Set Screw Reflectors

Elliptical Set Screw Reflectors

APPENDICES 197
**Set Screw Reflectors**

Carley Set Screw Reflectors are identical to our standard Aluminum reflectors except that they allow the customer to adjust the position of the lamp inside the reflector as often as desired. Another benefit of the set screw reflectors is that when a lamp burns out, only the lamp and lamp sleeve need to be replaced as the set screw reflector will last for years under normal conditions.

When working with a Set Screw Reflector, be sure that your light bulb has a protective metal sleeve around it. Click here for Lamp Sleeves.

<table>
<thead>
<tr>
<th>REFLECTOR</th>
<th>TYPE</th>
<th>DIA.</th>
<th>O.L.</th>
<th>REAM</th>
<th>REFOCUS</th>
<th>FOCAL</th>
<th>FOR LAMP</th>
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<tbody>
<tr>
<td>1925</td>
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<td>.252</td>
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</tr>
<tr>
<td>1933</td>
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<td>.50</td>
<td>.252</td>
<td>.05</td>
<td>T-1 1/2</td>
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</tr>
<tr>
<td>1934</td>
<td>Par</td>
<td>.500</td>
<td>.50</td>
<td>.189</td>
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<td>.252</td>
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<td>T-1 1/2</td>
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<tr>
<td>1935</td>
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<tr>
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<td>T-2 1/2</td>
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<tr>
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<td></td>
</tr>
</tbody>
</table>

* Par = Parabolic (Collimating) Ell = Elliptical (Refocusing).
** Optimum Refocus - refocus can be changed by moving lamp forward or backward inside the reflector.

---

**These reflectors can also be vacuum metalized after polishing, for increased reflectance - there is an additional charge for vacuum metalizing.**

---

**Lamp Sleeves/Reflector Sleeves**

The assemblies shown below are designed to fit Carley’s Set Screw Reflector Assemblies. Carley lamps size T-1, T-1 ½ and T-2 ½ can be used in any size reflector we offer. Be careful to order the correct sleeve for the lamp and reflector combination you will be combining.
Figure 1  T-1 Stiff Pin
Sleeve No. 1963

Figure 4 T-1 Wire Lead
Sleeve No. 1963

Figure 2  T-1 1/2 Stiff Pin
Sleeve No. 1964

Figure 5 T-1 1/2 Wire Lead
Sleeve No. 1964

Figure 3  T-2 1/2 Stiff Pin
Sleeve No. 1965

Figure 6 T-2 1/2 Wire Lead
Sleeve No. 1965

NOTE: Use sleeve 2129 for T-1 lamp to obtain T-2 ½ OD sleeve diameter.
Use sleeve 2139 for T-2 ½ lamp in 3 inch diameter reflector.
APPENDIX E

Data Sheet for Vertical Cavity Surface Emitting Laser

Model VC850-TO46-LM
850nm Oxide VCSEL TO-Can

**Features & Applications**
- 850 nm center optical wavelength
- Low dependence of electrical and optical characteristics over temperature
- Laser mouse

**Electro-Optical Specifications (T = 25 °C)**

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Symbol</th>
<th>Min.</th>
<th>Typ.</th>
<th>Max.</th>
<th>Unit</th>
<th>Test Condition</th>
</tr>
</thead>
<tbody>
<tr>
<td>Operating Voltage</td>
<td>$V_{op}$</td>
<td>***</td>
<td>1.9</td>
<td>2.2</td>
<td>V</td>
<td>$I=5mA$</td>
</tr>
<tr>
<td>Output Power</td>
<td>$P_0$</td>
<td>0.3</td>
<td>***</td>
<td>1</td>
<td>mW</td>
<td>$I=5mA$</td>
</tr>
<tr>
<td>Threshold Current</td>
<td>$I_{th}$</td>
<td>***</td>
<td>1</td>
<td>3</td>
<td>mA</td>
<td>*****</td>
</tr>
<tr>
<td>Slope Efficiency (S.E.)</td>
<td>$\eta$</td>
<td>0.1</td>
<td>0.25</td>
<td>0.35</td>
<td>mW/mA</td>
<td>$I=5mA$</td>
</tr>
<tr>
<td>Center Wavelength</td>
<td>$\lambda_c$</td>
<td>840</td>
<td>850</td>
<td>860</td>
<td>nm</td>
<td>$I=5mA$</td>
</tr>
<tr>
<td>Spectral Width (RMS)</td>
<td>$\Delta \lambda$</td>
<td>***</td>
<td>***</td>
<td>0.9</td>
<td>nm</td>
<td>$I=5mA$</td>
</tr>
<tr>
<td>Beam Divergence</td>
<td></td>
<td>20</td>
<td>***</td>
<td>10</td>
<td>15</td>
<td></td>
</tr>
<tr>
<td>Output Power Temp. Variation</td>
<td>$\Delta P/\Delta T$</td>
<td>-</td>
<td>-0.02</td>
<td>-</td>
<td>dBm/°C</td>
<td>25 to 85°C</td>
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</table>

**Package Dimensions (unit:mm)**

![Package Diagram](image)
Electro-Optical Characteristics Curves (25 °C)

Output Power vs. Forward Current

Forward Voltage vs. Forward Current
Output Power vs. Forward Current

At Different Ambient Temperature

Threshold Current vs. Ambient Temperature

E HIGHER WAY.

NO. 4, ROAD 10, INDUSTRIAL ZONE, TAICHUNG, TAIWAN

TEL: 04-23550011  FAX: 04-23550022

http://www.highervay.com.tw  E-MAIL: SERVICE@HIGHERVAY.COM.TW

APPENDICES
APPENDIX F

Data Sheet for High Speed Silicon Photodetector

Model S5821-03
Si PIN photodiode

S5821 series

High performance, high reliability Si PIN photodiodes

S5821 series is a high-speed Si PIN photodiode having high sensitivity over a wide spectral range from visible to near infrared light. S5821 series provides high performance and reliability at a low cost.

Features
- High-speed response
- Wide spectral response
- Low dark current
- Low terminal capacitance

Applications
- Optical switch
- Automobile optical sensor
- General photometry

General ratings / Absolute maximum ratings

<table>
<thead>
<tr>
<th>Type No.</th>
<th>Dimensional outline/ Window material</th>
<th>Package</th>
<th>Active area size</th>
<th>Effective active area</th>
<th>Reverse voltage</th>
<th>Power dissipation</th>
<th>Operating temperature</th>
<th>Storage temperature</th>
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</thead>
<tbody>
<tr>
<td>S5821</td>
<td>TO-18</td>
<td>ø1.2</td>
<td>1.1</td>
<td>20</td>
<td>50</td>
<td>-40 to +100</td>
<td>-55 to +125</td>
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</tr>
<tr>
<td>S5821-01</td>
<td>3/L</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>S5821-02</td>
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<td></td>
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<td></td>
<td></td>
</tr>
<tr>
<td>S5821-03</td>
<td>3/L</td>
<td></td>
<td></td>
<td></td>
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<td></td>
<td></td>
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Electrical and optical characteristics (Typ. Ta=25 °C, unless otherwise noted)

<table>
<thead>
<tr>
<th>Type No.</th>
<th>Spectral response range λp (nm)</th>
<th>Peak sensitivity λp S (AW)</th>
<th>Short circuit current at 100°C (µA)</th>
<th>Dark current to 100V (nA)</th>
<th>Temp. coefficient of λp to 10°C (times/°C)</th>
<th>Cutoff frequency at 10V (MHz)</th>
<th>Terminal capacitance at 10V f=1MHz (pF)</th>
<th>NEP (W/Hz^{1/2})</th>
</tr>
</thead>
<tbody>
<tr>
<td>S5821</td>
<td>200 to 1100</td>
<td>960</td>
<td>0.6</td>
<td>0.45</td>
<td>0.52</td>
<td>0.55</td>
<td>1.15</td>
<td>25</td>
</tr>
<tr>
<td>S5821-01</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>S5821-02</td>
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<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>S5821-03</td>
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<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

Window material K: borosilicate glass, L: lens type borosilicate glass
APPENDIX F

- Spectral response

![Spectral Response Graph](Typ. Ta=25 °C)

- Directivity

![Directivity Graph](Typ. Ta=25 °C)

- Photo sensitivity temperature characteristic

![Photo Sensitivity Temperature Graph](Typ.)

- Frequency response

![Frequency Response Graph](Typ. Ta=25 °C; L=50 mm, R=50 kΩ, V=10 V)

APPENDICES 208
Cut-off frequency vs. reverse voltage

Dark current vs. reverse voltage

Terminal capacitance vs. reverse voltage

(Si PIN photodiode, S5821 Series)
APPENDIX G

Hardware Specification for Discrete Sensing with Custom Electronics (DISCO4)
1 Introduction

This document describes the operation and use of the PPG Board developed by Cambridge Consultants Ltd. on behalf of Dialog Devices Ltd.

1.1 System Description

The PPG board implements a four-wavelength photoplethysmographic system that provides all the necessary analogue signal-processing to perform multi-channel photoplethysmography or pulse oximetry. Whilst the primary outputs of this board are analogue signals, the majority of system functions may be configured digitally, using an RS232 serial communications interface. In addition, the serial interface may be used to capture many of the signals and settings digitally. Jumpers and expansion connectors are provided to further enhance the applicability of the system to a range of physiological studies.
### 1.2 System Specifications

The PPG board is designed to work with a range of probes, both commercially available and custom. A summary of the major specifications follows:

<table>
<thead>
<tr>
<th>Specification</th>
<th>Details</th>
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<tbody>
<tr>
<td>Power Supply</td>
<td>±13V to ±18V</td>
</tr>
<tr>
<td>Number of Channels</td>
<td>4</td>
</tr>
<tr>
<td>Maximum LED drive current</td>
<td>100mA or 30mA - jumper selectable</td>
</tr>
<tr>
<td>Maximum LED forward voltage</td>
<td>5V</td>
</tr>
<tr>
<td>LED Configurations</td>
<td>Ground returned or back-to-back</td>
</tr>
<tr>
<td>LED drive-current resolution</td>
<td>8 bits over 30mA or 100mA range</td>
</tr>
<tr>
<td>LED current control</td>
<td>Manual (RS232) or Auto control</td>
</tr>
<tr>
<td>Transimpedance gain</td>
<td>1MV/A or 110kV/A - jumper selectable</td>
</tr>
<tr>
<td>Analogue Outputs</td>
<td>Raw + Processed</td>
</tr>
<tr>
<td>Raw Analogue Bandwidth</td>
<td>DC to 500Hz or 1KHz - jumper selectable</td>
</tr>
<tr>
<td>Raw Analogue Gain</td>
<td>6dB</td>
</tr>
<tr>
<td>Processed Analogue Bandwidth</td>
<td>0.75Hz to 11Hz or 400Hz - jumper selectable</td>
</tr>
<tr>
<td>Processing Filter Characteristics</td>
<td>3rd Order LPF + 5th Order HPF</td>
</tr>
<tr>
<td>Processed Analogue Gain</td>
<td>15dB to 77dB digitally controllable (+ 6dB fixed)</td>
</tr>
<tr>
<td>Analogue Gain Resolution</td>
<td>5 bits - linear dB scale (2dB per bit)</td>
</tr>
<tr>
<td>Serial Communications</td>
<td>Multi-drop addressed RS232</td>
</tr>
<tr>
<td>Switching Speed</td>
<td>1kHz / LED or 2kHz / LED - jumper selectable</td>
</tr>
<tr>
<td>Probe resistor measurement</td>
<td>100Ω to 5MΩ under serial-port control</td>
</tr>
<tr>
<td>Other connectors</td>
<td>Expansion Port, External preamplifier Connector, Microprocessor In-Circuit-Programming</td>
</tr>
<tr>
<td>PCB Dimensions</td>
<td>210mm x 250mm</td>
</tr>
</tbody>
</table>
2 Connectors

2.3 Probe Connector

The circuit uses a single 9-pin D-type female connector to connect to all probes. The pin-out, shown below, is chosen so that commercially available pulse-oximeter probes can be connected directly to the PCB. This pin-out is consistent with the Nellcor range of probes and anything which claims to be Nellcor compatible.

![9-pin D-Type Female Connector](attachment:image.png)

**Connections**

1 = PROBE_R  
2 = LED1  
3 = LED2  
4 = LED3  
5 = PIN_ANODE  
6 = GND_SCREEN  
7 = GND_SCREEN_PIN  
8 = LED4  
9 = PIN_CATHODE  

Custom probes can also be used with this connector, either directly or through an adapter cable. There are several options for connecting the probe components to the connector, which can all be used without changing jumper or software settings. The following sections describe the probe connector pins and the various options for wiring the probe components.

2.3.1 PROBE_R

A resistor can be connected from PROBE_R to either GND_SCREEN (preferred) or GND_SCREEN_PIN. The value of this resistor can then be measured by the PCB to identify the type of probe connected. This function is compatible with the Nellcor range of probes, which use resistors to validate that the probe is compatible with the instrument.

If this functionality is not required then PROBE_R can either be left open-circuit or can be connected to GND_SCREEN or GND_SCREEN_PIN.
2.3.2 LED1-4

The four LEDs can be connected to these pins in a variety of topologies, all of which can be used without configuration of the PCB. Each LED can either be connected from the relevant LED pin to GND SCREEN or to any other LED pin. This enables various combinations of ground-return and back-to-back configurations, as shown in the examples below.

The only restriction is that no more than one LED should be returned to any pin; this is equivalent to saying that two LEDs can be used back-to-back, but not three.

Each LED as shown in the examples can either be a single device or many connected in series. If series combinations are used then it should be ensured that the total forward voltage drop is less than 5V.

If any LED pin is not used then it should be left open-circuit.
2.4 Serial Port Connectors

The circuit uses two 9-pin D-type male connectors to implement the RS232 serial communications. The pin-outs of both MASTER and SLAVE connectors are identical and shown below:

![9-pin D-Type Male Connector Diagram]

**Connections**

1 = NC  
2 = RX - Received Data  
3 = TX - Transmitted Data  
4 = NC  
5 = GND  
6 = NC  
7 = NC  
8 = NC  
9 = NC

The MASTER connector is used to connect a single PCB to a PC, whilst the SLAVE connector is used to daisy-chain several PCBs together for connection to a single PC.

All connections between the PC and one or more PCBs should be made with standard female-to-female null-modem cables, as described in the following section.
2.5 Filtered Analogue Outputs

A single 16-way 0.1-inch pitch IDC connector is used as the output for all amplified and filtered output channels. The pin-out of this connector is shown below:

<table>
<thead>
<tr>
<th>Pin No.</th>
<th>Name</th>
<th>Description</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>CH1FSIG</td>
<td>Channel 1 Output Signal</td>
</tr>
<tr>
<td>2</td>
<td>GND</td>
<td>Signal Ground</td>
</tr>
<tr>
<td>3</td>
<td>CH2FSIG</td>
<td>Channel 2 Output Signal</td>
</tr>
<tr>
<td>4</td>
<td>GND</td>
<td>Signal Ground</td>
</tr>
<tr>
<td>5</td>
<td>CH3FSIG</td>
<td>Channel 3 Output Signal</td>
</tr>
<tr>
<td>6</td>
<td>GND</td>
<td>Signal Ground</td>
</tr>
<tr>
<td>7</td>
<td>CH4FSIG</td>
<td>Channel 4 Output Signal</td>
</tr>
<tr>
<td>8</td>
<td>GND</td>
<td>Signal Ground</td>
</tr>
<tr>
<td>9</td>
<td>GND</td>
<td>Signal Ground</td>
</tr>
<tr>
<td>10</td>
<td>GND</td>
<td>Signal Ground</td>
</tr>
<tr>
<td>11</td>
<td>VDD</td>
<td>+ve Power Supply</td>
</tr>
<tr>
<td>12</td>
<td>GND</td>
<td>Signal Ground</td>
</tr>
<tr>
<td>13</td>
<td>-VDD</td>
<td>-ve Power Supply</td>
</tr>
<tr>
<td>14</td>
<td>GND</td>
<td>Signal Ground</td>
</tr>
<tr>
<td>15</td>
<td>GND</td>
<td>Signal Ground</td>
</tr>
<tr>
<td>16</td>
<td>GND</td>
<td>Signal Ground</td>
</tr>
</tbody>
</table>

These outputs are the main analogue outputs of the entire system and are taken from the very end of the analogue signal-processing chain.

These outputs should not be loaded by an impedance of less than 10kΩ without external buffering. The power supplies must not be used to supply more than 50mA to external circuitry.
2.6 Raw Outputs

A single 16-way 0.1-inch pitch IDC connector is used as the output for the unprocessed output channels. The pin-out of this connector is shown below:

<table>
<thead>
<tr>
<th>Pin No.</th>
<th>Name</th>
<th>Description</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>CH1SIG</td>
<td>Channel 1 Raw Output Signal</td>
</tr>
<tr>
<td>2</td>
<td>GND</td>
<td>Signal Ground</td>
</tr>
<tr>
<td>3</td>
<td>CH2SIG</td>
<td>Channel 2 Raw Output Signal</td>
</tr>
<tr>
<td>4</td>
<td>GND</td>
<td>Signal Ground</td>
</tr>
<tr>
<td>5</td>
<td>CH3SIG</td>
<td>Channel 3 Raw Output Signal</td>
</tr>
<tr>
<td>6</td>
<td>GND</td>
<td>Signal Ground</td>
</tr>
<tr>
<td>7</td>
<td>CH4SIG</td>
<td>Channel 4 Raw Output Signal</td>
</tr>
<tr>
<td>8</td>
<td>GND</td>
<td>Signal Ground</td>
</tr>
<tr>
<td>9</td>
<td>GND</td>
<td>Signal Ground</td>
</tr>
<tr>
<td>10</td>
<td>GND</td>
<td>Signal Ground</td>
</tr>
<tr>
<td>11</td>
<td>VDD</td>
<td>+ve Power Supply</td>
</tr>
<tr>
<td>12</td>
<td>GND</td>
<td>Signal Ground</td>
</tr>
<tr>
<td>13</td>
<td>-VDD</td>
<td>-ve Power Supply</td>
</tr>
<tr>
<td>14</td>
<td>GND</td>
<td>Signal Ground</td>
</tr>
<tr>
<td>15</td>
<td>SIG_IN</td>
<td>Amplifier Output Signal before De-Multiplexing</td>
</tr>
<tr>
<td>16</td>
<td>GND</td>
<td>Signal Ground</td>
</tr>
</tbody>
</table>

The raw output signals are the various channels after de-multiplexing but before any analogue signal-processing (amplification and gain) have been applied. This connector is provided so that external analogue signal-processing could be used if required.

These outputs should not be loaded by an impedance of less than 10kΩ without external buffering. The power supplies must not be used to supply more than 50mA to external circuitry.
APPENDIX H

Hardware Specification for Labjack U12
often interfere with the installation of the LVRTE. If you have trouble running the example applications, repeat the LabJack software installation to make sure the LVRTE is installed.

To test the installation, start LJtest by selecting

*Start* => *Programs* => *LabJack* => *LJtest.*

Make sure "Test Fixture Installed" and "Continuous" are not selected, and press the "Run" button. LJtest will step through 8 separate tests and all should pass.
2. Hardware Description

The external features of the LabJack U12 are:
- USB connector,
- DB25 digital I/O connector,
- Status LED,
- 30 screw terminals.

The USB connection provides power and communication. No external power supply is needed. The +5 volt connections available at various locations are outputs, do not connect a power supply.

Figure 2-1 shows the top surface of the LabJack U12. Not shown is the USB and DB25 connector, which are both on the top edge. The DB25 connector provides connections for 16 digital I/O lines, called D0-D15. It also has connections for ground and +5 volts. All connections besides D0-D15, are provided by the 30 screw terminals shown in Figure 1. Each individual screw terminal has a label, A10 through STB.

The status LED blinks 4 times at power-up, and then blinks once and stays on after enumeration (recognition of the LabJack U12 by the PC operating system). The LED also blinks during burst and stream operations, unless disabled. The LED can be enabled/disabled through software using the functions AISample, AIBurst, or AIStreamStart. Since the LED uses 4-5 mA of current, some users might wish to disable it for power-sensitive applications.
2.1 A10 – A17

Hardware
The LabJack U12 has 8 screw terminals for analog input signals. These can be configured individually and on-the-fly as 8 single-ended channels, 4 differential channels, or combinations in between. Each input has a 12-bit resolution and an input bias current of ±90 μA.

- Single-Ended: The input range for a single-ended measurement is ±10 volts.
- Differential channels can make use of the low noise precision PGA to provide gains up to 20. In differential mode, the voltage of each AI with respect to ground must be between +20 and -10 volts, but the range of voltage difference between the 2 AI is a function of gain (G) as follows:

<table>
<thead>
<tr>
<th>G</th>
<th>Voltage Difference</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>±20 volts</td>
</tr>
<tr>
<td>2</td>
<td>±10 volts</td>
</tr>
<tr>
<td>4</td>
<td>±5 volts</td>
</tr>
<tr>
<td>5</td>
<td>±4 volts</td>
</tr>
<tr>
<td>8</td>
<td>±2.5 volts</td>
</tr>
<tr>
<td>10</td>
<td>±2 volts</td>
</tr>
<tr>
<td>16</td>
<td>±1.25 volts</td>
</tr>
<tr>
<td>20</td>
<td>±1 volt</td>
</tr>
</tbody>
</table>

The reason the range is ±20 volts at G=1 is that, for example, A10 could be +10 volts and A11 could be -10 volts giving a difference of ±20 volts, or A10 could be -10 volts and A11 could be +10 volts giving a difference of -20 volts.

The PGA (programmable gain amplifier, available on differential channels only) amplifies the AI voltage before it is digitized by the A/D converter. The high level drivers then divide the reading by the gain and return the actual measured voltage.

Figure 2-2 shows a typical single-ended connection measuring the voltage of a battery. This same measurement could also be performed with a differential connection to allow the use of the PGA. In general, any single-ended measurement can be performed using a differential channel by connecting the voltage to an even-numbered analog input, and grounding the associated odd-numbered analog input (as shown by the dashed connection to A11 in Figure 2-2).

![Figure 2-2. Single-ended measurement.](image)

Figure 2-3 shows a typical differential connection measuring the voltage across a current shunt. A differential connection is required when neither leg of the shunt is at ground potential. Make sure that the voltage of both A10 and A11 with respect to ground is within ±10 volts. For instance, if the source (Vs) shown in Figure 2-3 is 120 VAC, the difference between A10 and A11 might be
small, but the voltage from both AI0 and AI1 to ground will have a maximum value near 170 volts, and will seriously damage the LabJack.

Whether or not the ground (GND) connection is needed (Figure 2-3) will depend on the nature of Vs.

\[ R_2 = \frac{V_a R_1}{(V_s - V_a)} \]

Figure 2-3 shows a single-ended connection used to measure the output voltage of a typical voltage-divider circuit. The voltage divider circuit is a simple way to convert a varying resistance (thermistor, photoresistor, potentiometer, etc.) to a varying voltage. With nothing connected to Va, the value of the unknown resistance, R2, can be calculated as:

\[ R_2 = V_a R_1 / (V_s - V_a) \]

where Vs is the supply voltage (+5V in Figure 2-4).

When Va is connected to AI0, as shown in Figure 2-4, the input bias current of the LabJack affects the voltage divider circuit, and if the resistance of R1 and R2 is too large, this effect must be accounted for or eliminated. This is true for any signal with too high of a source impedance.

All measuring devices have maximum analog input bias currents that vary from picoamps to milliamps. The input bias current of the LabJack U12's analog inputs varies from +70 to -94 microamps (μA). This is similar to an input impedance of about 100 kΩ, but because the current is nonzero at 0 volts, it is better to model the analog input as a current sink obeying the following rule:

\[ I_{in} = 8.181 V_a - 11.67 \, \mu A \]
Figure 2-4. Single-ended measurement with voltage divider circuit.

Because the input bias current is known, as a function of input voltage, the simple voltage divider equation can be modified as follows to account for input bias current:

\[ R_2 = \frac{V_a}{\left[\frac{(V_s-V_a)}{R_1}\right] - (8.181 \mu A \cdot V_a) + 11.67 \mu A} \]

As an alternative to the equation above, \( V_a \) can be buffered by a single-supply rail-to-rail operational amplifier, and the original simple voltage divider equation can be used. This solution works for any single-ended signal which stays between 0 and +5 volts. Some op-amp choices are:

- TLV2462
- LMC6482
- MAX4166

Software
Readings from the analog inputs are returned by the functions EAnalogIn, AISample, AI Burst, and AI StreamRead.

EAnalogIn is a simplified (E is for easy) function that returns a single reading from 1 analog input channel. Execution time is up to 20 ms.

AISample returns a single reading of 1-4 channels, and takes up to 20 ms to execute, providing a maximum date rate of about 50 Hz per channel.

AI Burst acquires multiple samples of 1-4 channels at a hardware-timed sample rate of 400-8192 Hz. The acquisition can be triggered based on a change of state on IO0 or IO1. This function also returns the states of the IO pins (which are read every 4 samples).

Internally, the actual number of samples collected and transferred by the LabJack during an AI Burst call is the smallest power of 2, from 64 to 4096, which is at least as big as numSamples. The execution time of this function, in milliseconds, can be estimated as:

- Turbo (default) => 30+(1000*numSamplesActual/sampleRate)+(0.4*numSamplesActual)
- Normal => 30+(1000*numSamplesActual/sampleRate)+(2.5*numSamplesActual)

numSamples = numScans * numChannels
sampleRate = scanRate * numChannels
AIStreamRead is called periodically during a stream acquisition started by AIStreamStart. Each call retrieves multiple samples of 1-4 channels from the LabJack stream buffer, along with the states of the IO pins (read every 4 samples). Hardware-timed sample rates of 200-1200 Hz are available. If any function besides AIStreamRead is called while a stream is in progress, the stream will be stopped.

2.2 A00 & A01
The LabJack U12 has 2 screw terminals for analog output voltages. Each analog output can be set to a voltage between 0 and the supply voltage (+5 volts nominal) with 10-bits of resolution.

The output voltage is ratiometric with the +5 volt supply (+5V), which is generally accurate to ±5% (see Appendix A). If an output voltage of 5 volts is specified, the resulting output will be 100% of the supply voltage. Similarly, specifying 2.5 volts actually gives 50% of the supply voltage. The maximum output voltage is almost 100% of +5V at no-load, and decreases with load. See the specifications in Appendix A relating to maximum output voltage. Also note that loading either analog output will cause an IR drop through the source impedance of each.

If improved accuracy is needed, measure the +5 volt supply with an analog input channel, and the actual output voltage can be calculated. For instance, if an analog output of 2.5 volts is specified and a measurement of +5V returns 5.10 volts, the actual output voltage is 2.55 volts (at no-load). Alternatively (and preferably), the analog output can itself be measured with an analog input.

There is a 1st order low-pass filter on each analog output with a 3dB frequency around 22 Hz.

The analog outputs are initialized to 0.0 volts on power-up or reset.

The analog outputs can withstand a continuous short-circuit to ground, even when set at maximum output.

Voltage should never be applied to the analog outputs, as they are voltage sources themselves. In the event that a voltage is accidentally applied to either analog output, they do have protection against transient overvoltages such as ESD (electrostatic discharge) and continuous overvoltage of a couple volts. An applied voltage that exceeds the capability of this protection will most likely damage the resistor R63 (A00) or R62 (A01) on the LabJack U12 PCB. The symptom of such a failure would be reduced voltage from the analog outputs, particularly at load, and could be verified by measuring the resistance of R62/R63 (should be less than 50 ohms but a damaged resistor will measure higher). A simple repair for such damage is to remove the damaged resistor and simply make a short with a blob of solder.

Software
The analog outputs are set using the function EAnalogOut (easy function) or AOUpdate, which take up to 20 ms to execute, providing a maximum update rate of about 50 Hz per channel. AOUpdate also controls/reads all 20 digital I/O and the counter.

2.3 I00 – I03
Connections to 4 of the LabJack’s 20 digital I/O are made at the screw terminals, and are referred to as I00-I03. Each pin can individually be set to input, output high, or output low. These 4 channels include a 1.5 kΩ series resistor that provides overvoltage/short-circuit protection. Each channel also has a 1 MΩ resistor connected to ground.
All digital I/O are set to input on power-up or reset.

One common use of a digital input is for measuring the state of a switch as shown in Figure 2-5. If the switch is open, IO0 reads FALSE. If the switch is closed, IO0 reads TRUE.

![Figure 2-5. IO used to detect the state of a switch.](image)

While providing overvoltage/short-circuit protection, the 1.5 kΩ series resistor on each IO pin also limits the output current capability. For instance, with an output current of 1 mA, the series resistor will drop 1.5 volts, resulting in an output voltage of about 3.5 volts.

**Software**

The easy functions EDigitalIn or EDigitalOut are used to read or set the state of one digital line, and both take up to 20 ms to execute.

The functions AOUpdate and DigitalIO are used to set the direction, set the state, and/or read the state, of each IO pin. Both of these functions take up to 20 ms to execute, providing a maximum update rate of about 50 Hz per pin.

The function AISample can set/read the state of each IO, but setting the state will have no effect unless the IO have been configured as outputs using another function. The function Counter reads the state of each IO.

The functions AI Burst and AI Stream Read, take a reading of the IO states and return it with the analog data. The states of the 4 IO are read simultaneously every 4 samples, providing a data rate of up to 2048 Hz per pin for burst mode, or 300 Hz per pin for stream mode. For 1 or 2 channel scans, duplicate data (4x or 2x) will be added to the read array such that the size is numScans.

2.4 **D0 – D15**

Connections to 16 of the LabJack's 20 digital I/O are made at the DB25 connector, and are referred to as D0-D15. These 16 lines have no overvoltage/short-circuit protection, and can sink or source up to 25 mA each (total sink or source current of 200 mA max for all 16). This allows the D pins to be used to directly control some relays. All digital I/O are CMOS output and TTL input except for D13-D15, which are Schmitt trigger input. Each D pin has a 1 MΩ resistor connected to ground.

All digital I/O are set to input on power-up or reset.
APPENDIX H

DB25 Pinouts:

<table>
<thead>
<tr>
<th>Pin</th>
<th>Connection</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>D0</td>
</tr>
<tr>
<td>2</td>
<td>D1</td>
</tr>
<tr>
<td>3</td>
<td>D2</td>
</tr>
<tr>
<td>4</td>
<td>D3</td>
</tr>
<tr>
<td>5</td>
<td>D4</td>
</tr>
<tr>
<td>6</td>
<td>D5</td>
</tr>
<tr>
<td>7</td>
<td>D6</td>
</tr>
<tr>
<td>8</td>
<td>D7</td>
</tr>
<tr>
<td>9</td>
<td>NC</td>
</tr>
<tr>
<td>10</td>
<td>+5V</td>
</tr>
<tr>
<td>11</td>
<td>+5V</td>
</tr>
<tr>
<td>12</td>
<td>+5V</td>
</tr>
<tr>
<td>13</td>
<td>+5V</td>
</tr>
<tr>
<td>14</td>
<td>GND</td>
</tr>
<tr>
<td>15</td>
<td>GND</td>
</tr>
<tr>
<td>16</td>
<td>GND</td>
</tr>
<tr>
<td>17</td>
<td>GND</td>
</tr>
<tr>
<td>18</td>
<td>D8</td>
</tr>
<tr>
<td>19</td>
<td>D9</td>
</tr>
<tr>
<td>20</td>
<td>D10</td>
</tr>
<tr>
<td>21</td>
<td>D11</td>
</tr>
<tr>
<td>22</td>
<td>D12</td>
</tr>
<tr>
<td>23</td>
<td>D13</td>
</tr>
<tr>
<td>24</td>
<td>D14</td>
</tr>
<tr>
<td>25</td>
<td>015</td>
</tr>
</tbody>
</table>

These digital I/O can detect the state of a switch using the same circuit shown in Figure 2-5.

Because the D pins have no overvoltage/short-circuit protection, the user must be careful to avoid damage. A series resistor can provide substantial protection for these pins (see the CB25 datasheet). The following are examples of things that could damage a D pin and/or the entire LabJack:

- Shorting a high output to ground (or any potential other than +5V).
- Shorting a low output to a nonzero voltage (such as +5V).
- Exceeding the voltage limits specified in Appendix A.

Software
The easy functions EDigitalIn or EDigitalOut are used to read or set the state of one digital line, and both take up to 20 ms to execute.

The functions AOUpdate and DigitalIO are used to set the direction, set the state, and/or read the state, of each D pin. In addition, DigitalIO also returns the current state of the direction and output registers. Both of these functions take up to 20 ms to execute, providing a maximum update rate of about 50 Hz per pin.

2.5 CNT
The input connection to the 32-bit counter is made at screw-terminal CNT. The counter is incremented when it detects a falling edge followed by a rising edge. This means that if you reset the counter while your signal is low, you will not get the first count until it goes high-low-high. In situations where this first count is important, you should simply substract the initial count from the final count, rather than doing a reset.

Software
The functions ECount (easy function), AOUpdate, and Counter are used to reset or read the counter. If a reset is specified, the counter is read first. All of these functions take up to 20 ms to execute, providing a maximum update rate of about 50 Hz.

Counter readings can also be returned in stream mode (AIStreamRead) at up to 300 Hz.

2.6 CAL – STB
These terminals are used during testing and calibration. CAL is a precision 2.5 volt reference, and can be used during normal operation, but care should be taken to observe the current limits specified in Appendix A. The CAL pin is protected from ESD and overvoltage, but severe overvoltage (steady-state or transient) can damage CAL, and result in the failure of all analog inputs.

2.7 +5V
The LabJack has a nominal +5 volt internal power supply. Power can be drawn from this power supply by connecting to the +5V screw-terminals, or the +5V pins on the DB25 connector. The
total amount of current that can be drawn from the +5V pins, analog outputs, and digital outputs, is 450 mA for most desktop computers and self-powered USB hubs. Some notebook computers and bus-powered hubs will limit this available current to about 50 mA.

The USB specification requires all hosts and hubs to have overcurrent protection. If the user puts too large a load on +5V (including a short circuit of +5V to GND) of the LabJack U12 (a USB device), the host or hub is responsible for limiting the current.

2.8 GND

The GND connections available at the screw-terminals and DB25 connector provide a common ground for all LabJack functions. They are all the same.

Caution should be used whenever making connections with systems that have their own power source. It is normal to connect U12 ground to other grounds to create a common reference, but the risk is that the U12 ground will become the preferred ground for the other systems and they could try to send high currents into the U12. To prevent this it is often a good idea to put a 10-100 ohm resistor (or even a fuse) in series with GND on the U12 and any grounds from active systems.

2.9 OEM Versions

The LabJack U12 is also available in 2 OEM (original equipment manufacturer) versions:

- **LJU12-PH**: This is a populated LabJack U12 PCB with pin-headers installed (on the component side of the PCB) instead of screw-terminals. Also, the LED is installed on the component side of the PCB, so nothing is installed on the solder side.
- **LJU12-NTH**: This is a populated LabJack U12 PCB with no through-hole components (DB25 connector, USB connector, LED, screw-terminals). This board is meant for OEMs who solder connections directly to the PCB, or wish to install only certain connectors.

Dimensional drawings are available from the downloads page at labjack.com.

Normally, nothing ships with these OEM LabJacks except for the populated PCB. All software is of course available online at labjack.com.
APPENDIX I

Matlab Flow Chart and Source Code
### A. Specifications

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Conditions</th>
<th>Min</th>
<th>Typical</th>
<th>Max</th>
<th>Units</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>General</strong></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>USB Cable Length</td>
<td>CE compliance</td>
<td></td>
<td></td>
<td>20</td>
<td>meters</td>
</tr>
<tr>
<td>User Connection(s) Length</td>
<td></td>
<td></td>
<td></td>
<td>3</td>
<td>meters</td>
</tr>
<tr>
<td>Supply Current (1)</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Operating Temperature</td>
<td>-25 °C</td>
<td></td>
<td></td>
<td>85</td>
<td>°C</td>
</tr>
<tr>
<td>Clock Error</td>
<td>-40 to 70 °C</td>
<td></td>
<td></td>
<td>±30</td>
<td>ppm</td>
</tr>
<tr>
<td></td>
<td>-40 to 85 °C</td>
<td></td>
<td></td>
<td>±50</td>
<td>ppm</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td></td>
<td>±100</td>
<td>ppm</td>
</tr>
<tr>
<td><strong>+5 Volt Power Supply (+5V)</strong></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Voltage (Vs) (2)</td>
<td>Self-Powered</td>
<td>4.5</td>
<td></td>
<td>5.25</td>
<td>volts</td>
</tr>
<tr>
<td></td>
<td>Bus-Powered</td>
<td>4.1</td>
<td></td>
<td>5.25</td>
<td>volts</td>
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<tr>
<td>Output Current (2) (3)</td>
<td>Self-Powered</td>
<td>450</td>
<td></td>
<td>500</td>
<td>mA</td>
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<tr>
<td></td>
<td>Bus-Powered</td>
<td>50</td>
<td></td>
<td>100</td>
<td>mA</td>
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<tr>
<td><strong>Analog Inputs (A10 - A17)</strong></td>
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<tr>
<td>Input Range For Linear Operation</td>
<td>A1x to GND, SE</td>
<td>-10</td>
<td></td>
<td>10</td>
<td>volts</td>
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<tr>
<td></td>
<td>A1x to GND, Diff.</td>
<td>-10</td>
<td></td>
<td>20</td>
<td>volts</td>
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<tr>
<td>Maximum Input Range</td>
<td>A1x to GND</td>
<td>-40</td>
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<td>40</td>
<td>volts</td>
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<tr>
<td>Input Current (4)</td>
<td>Vin = +10 volts</td>
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<td>Vin = 0 volts</td>
<td>-11.7</td>
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<td>µA</td>
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<tr>
<td></td>
<td>Vin = -10 volts</td>
<td>-93.5</td>
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<td>µA</td>
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<tr>
<td>Resolution (No Missing Codes)</td>
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<td>Burst Diff. (5)</td>
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<tr>
<td></td>
<td>Burst SE (5)</td>
<td>11</td>
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<td>bits</td>
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<tr>
<td>Offset</td>
<td>G = 1 to 20</td>
<td>±1 G</td>
<td></td>
<td>bits</td>
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<td>Absolute Accuracy</td>
<td>SE</td>
<td>±0.2</td>
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<td>% FS</td>
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<tr>
<td></td>
<td>Diff.</td>
<td>±1</td>
<td></td>
<td>% FS</td>
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<td>Noise</td>
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<td>Differential Linearity Error</td>
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<td>Repeatability</td>
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<td>±0.05</td>
<td>±0.25</td>
<td>%</td>
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<td>Sink</td>
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<td>100</td>
<td>µA</td>
<td></td>
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<td>Burst</td>
<td>25</td>
<td>50</td>
<td>µs</td>
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</tr>
<tr>
<td>Trigger Latency</td>
<td>Burst</td>
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<td>µs</td>
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<td>Analog Outputs (AO0 &amp; AO1)</td>
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<tr>
<td>Maximum Voltage (6)</td>
<td>No Load</td>
<td>Vs</td>
<td></td>
<td></td>
<td>volts</td>
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<tr>
<td></td>
<td>At 1 mA</td>
<td>0.99 * Vs</td>
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<td>volts</td>
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<tr>
<td></td>
<td>At 5 mA</td>
<td>0.96 * Vs</td>
<td></td>
<td>volts</td>
<td></td>
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<td>Source Impedance</td>
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<td></td>
<td></td>
<td>Ω</td>
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<td>Output Current</td>
<td>Each AO</td>
<td></td>
<td></td>
<td>20</td>
<td>mA</td>
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APPENDICES 230
### APPENDIX H

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Conditions</th>
<th>Min</th>
<th>Typical</th>
<th>Max</th>
<th>Units</th>
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<tr>
<td><strong>IO</strong></td>
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<td>Low Level Input Voltage</td>
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<td>3</td>
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<td>High Level Input Voltage</td>
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<td>volts</td>
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<td>Input Leakage Current (7)</td>
<td>Output High</td>
<td>3.3</td>
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<td>mA</td>
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<td>Output Short-Circuit Current (8)</td>
<td>No Load</td>
<td>Vs</td>
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<td>volts</td>
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<tr>
<td>Output Voltage (8)</td>
<td>At 1 mA</td>
<td>Vs</td>
<td></td>
<td></td>
<td>volts</td>
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<td></td>
<td></td>
<td>Vs - 0.4</td>
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<td>Vs - 1.5</td>
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<td>volts</td>
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<tr>
<td>Low Level Input Voltage (9)</td>
<td>D0 - D12</td>
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<td>D13 - D15</td>
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<td>D0 - D12</td>
<td>Vs + 0.3</td>
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<td>volts</td>
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<tr>
<td></td>
<td>D13 - D15</td>
<td>Vs + 0.3</td>
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<td>volts</td>
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<tr>
<td>Input Leakage Current (7)</td>
<td>Per Line</td>
<td>±1</td>
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<td></td>
<td>µA</td>
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<td>Output Current (9)</td>
<td>Total D0 - D15</td>
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<td>mA</td>
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<td>Output Low Voltage</td>
<td>Vs - 0.7</td>
<td>0.6</td>
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<tr>
<td>Output High Voltage</td>
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<td><strong>CNT</strong></td>
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<td>Low Voltage (10)</td>
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<td>volts</td>
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<tr>
<td>High Voltage (10)</td>
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<td>15</td>
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<td></td>
<td>volts</td>
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<tr>
<td>Schmitt Trigger Hysteresis</td>
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<td>20-100</td>
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<td></td>
<td>mV</td>
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<tr>
<td>Input Leakage Current (7)</td>
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<td>±1</td>
<td></td>
<td></td>
<td>µA</td>
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<td>Minimum High Time</td>
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<td>ns</td>
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<tr>
<td>Maximum Input Frequency</td>
<td></td>
<td>1</td>
<td></td>
<td></td>
<td>MHz</td>
</tr>
</tbody>
</table>

(1) Current drawn by the LabJack through the USB. The status LED is responsible for 4-5 mA of this current.

(2) Self-powered would apply to USB hubs with a power supply, all known desktop computer USB hosts, and some notebook computer USB hosts. Bus-powered would apply to USB hubs without a power supply and some notebook computer USB hosts.

(3) This is the total current that can be sourced by +5V, analog outputs, and digital outputs.

(4) The input current at each analog input is a function of the voltage at that input (Vin) with respect to ground and can be calculated as: (8.1B1*Vin - 11.67)mA.

(5) Single-ended burst mode only returns even binary codes, and thus has a net resolution of 11 bits. In addition, extra noise in burst mode can reduce the effective resolution.

(6) Maximum analog output voltage is equal to the supply voltage at no load.

(7) Must also consider current due to 1 MΩ resistor to ground.

(8) The IO lines each have a 1500 ohm series resistor.

(9) These lines have no series resistor. It is up to the user to make sure the maximum voltages and currents are not exceeded.

(10) CNT is a Schmitt Trigger input.
Function ParseStream

START

Input: DISCO file

Prompt for filename

Filename specified?

Yes

Open file

File open successful?

Yes

Initialize variables

Found header flag?

Yes

Identify last line of header

Initialize start

End of header?

No

Compute number of column

Yes

Concatenate header

Line number > 10?

Yes

Compute maximum size of header

Size of header < 7?

No

Decrement size of header

Yes

Found string 'Comment'?

No

Set found header flag

Yes

Increment line number

Header string = Delta_x?

Yes

Line number > 10?

No

Delta = line number

Increment line number

Output: File in Matlab format

STOP

APPENDIX I

APPENDICES 233
% Filename: ParseStream
% A function to convert text files from DISCO4 into Matlab format.
% Input: DISCO4 file, usually a text file
% Output: Data in Matlab format

function [data,header]=ParseStream(fname)

% Prompt for filename if not specified
if nargin == 0
    fname = input('Data filename: ','s')
end

fid = fopen(fname,'rt'); % open file for reading, text mode

% Check if the file is open successfully
if fid == -1
    fprintf(2,'Cannot open file %s for reading, Quitting
',fname);
    data = 0; % Quit Flag
    return
end

% Empty string
header = "";
sift = "%"
found = 0;
n = 1;
delta = 0;
column = 0;

% Chomp through header
while found == 0 && n<30
    header = strvcat(header,fgetl(fid));
    if n > 10
        switch delta
            case (0)
                if strcmp ('Delta_X',header(n,1:7)) %find line that starts with Delta_X
                    delta = n;
                end
            otherwise
                end
        end
    end
    if n > 10
        for m = max(size(header)):1:7 %find last word in header
            if strcmp ('Comment',header(n,m-6:m))
                found = 1; %stop putting data in header when found
            end
        end
        n = n+1;
    end
end
% Detect number of columns
sift = header(delta,1:max(size(header))); 

for n = 1:max(size(header))
    column = column + strcmp(';',sift(n));
end

column = column + 1;

% Read remaining data
data = fscanf(fid,'%f
',[column,int]);

fclose(fid);    % close file

return
Main program *AnalyseNormSignal*

START

- Load Matlab data file

- Obtain coefficients for 10th order low pass Butterworth filter

- Filter AC and DC signal using the coefficients

- Identify start point for each offset positions

- Identify AC signal

- Identify DC signal

- Normalise Signal

- Completed all experiment data sets?

- Initialise start of offset positions

END of offset positions?

- Calculate mean normalised AC across the offset positions

- Calculate standard deviation for the normalised mean AC

- Calculate mean normalised DC across the offset positions

- Calculate standard deviation for the normalised mean DC

- Plot Errorbar graph for normalised AC and DC

- No

- Calculate sample of AC PPG signal for all experiment data sets

- Compute autocorrelation coefficient and lag time for all experiment data sets

- Plot autocorrelation graph for all experiment data sets

- STOP

APPENDICES
APPENDIX I

% Filename: AnalyseNormSignal
% To analyse the normalised AC and DC mean amplitude, plot the AC PPG signal at each offset positions
% and to plot the autocorrelation graph for the AC signal.
% This main calling program will be repeated all the experiments performed on gripped fingers and fingers
% with inter-digit gaps for both contact and non-contact transmission and reflection probes.
% This Matlab code shows the example of analysing PPG signal recorded on gripped fingers.

% Data Set 1
load gripTx01 % Load Matlab file on gripped fingers
[b,a] = butter(10,0.1); % Obtain coefficient for Butterworth filter
filterSig01 = filter(b,a,gripTx01); % Filter PPG signal
startPt = [4400 15000 25500 36500 43000 52500 65000 74000 86000];
ACsig = -filterSig01(:,4); % Convert AC from intensity measurement to volume measurement
DCsig = filterSig01(:,2);
[grip(1).AC,grip(1).DC,grip(1).meanDC,grip(1).meanAC] = NormaliseSignal(ACsig, DCsig, startPt);

% Data Set 2
load gripTx02
[b,a] = butter(10,0.1);
filterSig02 = filter(b,a,gripTx02);
startPt = [7000 14000 27000 38000 48000 58000 68000 79000 89000];
ACsig = -filterSig02(:,4);
DCsig = filterSig02(:,2);
[grip(2).AC,grip(2).DC] = NormaliseSignal(ACsig, DCsig, startPt);

% Data Set 3
load gripTx03
[b,a] = butter(10,0.1);
filterSig03 = filter(b,a,gripTx03);
startPt = [5500 15000 27000 37000 48000 57500 63000 82500 92500];
ACsig = -filterSig03(:,4);
DCsig = filterSig03(:,2);
[grip(3).AC,grip(3).DC] = NormaliseSignal(ACsig, DCsig, startPt);

% Data Set 4
load gripTx04
[b,a] = butter(10,0.1);
filterSig04 = filter(b,a,gripTx04);
startPt = [4000 15000 26500 36000 44000 56000 66000 71500 88000];
ACsig = -filterSig04(:,4);
DCsig = filterSig04(:,2);
[grip(4).AC,grip(4).DC] = NormaliseSignal(ACsig, DCsig, startPt);

% Data Set 5
load gripTx05
[b,a] = butter(10,0.1);
filterSig05 = filter(b,a,gripTx05);
startPt = [6000 16000 23000 35000 47000 55000 66000 79000 89000];
ACsig = -filterSig05(:,4);
DCsig = filterSig05(:,2);
[grip(5).AC,grip(5).DC] = NormaliseSignal(ACsig, DCsig, startPt);

% Data Set 6
load gripTx06
\[ [b,a] = \text{butter}(10,0.1); \]
\[ \text{filterSig06} = \text{filter}(b,a,\text{gripTx06}); \]
\[ \text{startPt} = [5000 \ 15000 \ 25000 \ 35500 \ 46000 \ 56000 \ 66000 \ 76000 \ 83000]; \]
\[ \text{ACsig} = \text{filterSig06}(:,4); \]
\[ \text{DCsig} = \text{filterSig06}(:,2); \]
\[ [\text{grip}(6).\text{AC},\text{grip}(6).\text{DC}] = \text{NormaliseSignal} (\text{ACsig}, \text{DCsig}, \text{startPt}); \]

% Data Set 7
load gripTx07
\[ [b,a] = \text{butter}(10,0.1); \]
\[ \text{filterSig07} = \text{filter}(b,a,\text{gripTx07}); \]
\[ \text{startPt} = [6500 \ 15500 \ 27000 \ 35500 \ 43500 \ 58000 \ 67000 \ 75000 \ 88000]; \]
\[ \text{ACsig} = \text{filterSig07}(:,4); \]
\[ \text{DCsig} = \text{filterSig07}( :,2); \]
\[ [\text{grip}(7).\text{AC},\text{grip}(7).\text{DC}] = \text{NormaliseSignal} (\text{ACsig}, \text{DCsig}, \text{startPt}); \]

% Data 8
load gripTx08
\[ [b,a] = \text{butter}(10,0.1); \]
\[ \text{filterSig08} = \text{filter}(b,a,\text{gripTx08}); \]
\[ \text{startPt} = [2000 \ 13000 \ 23000 \ 38000 \ 47500 \ 55500 \ 62000 \ 76000 \ 84000]; \]
\[ \text{ACsig} = \text{filterSig08}(:,4); \]
\[ \text{DCsig} = \text{filterSig08}( :,2); \]
\[ [\text{grip}(8).\text{AC},\text{grip}(8).\text{DC}] = \text{NormaliseSignal} (\text{ACsig}, \text{DCsig}, \text{startPt}); \]

% Data 9
load gripTx09
\[ [b,a] = \text{butter}(10,0.1); \]
\[ \text{filterSig09} = \text{filter}(b,a,\text{gripTx09}); \]
\[ \text{startPt} = [9000 \ 19000 \ 27500 \ 38000 \ 48000 \ 57500 \ 68000 \ 75500 \ 90000]; \]
\[ \text{ACsig} = \text{filterSig09}( :,4); \]
\[ \text{DCsig} = \text{filterSig09}( :,2); \]
\[ [\text{grip}(9).\text{AC},\text{grip}(9).\text{DC}] = \text{NormaliseSignal} (\text{ACsig}, \text{DCsig}, \text{startPt}); \]

% Data 10
load gripTx10
\[ [b,a] = \text{butter}(10,0.1); \]
\[ \text{filterSig10} = \text{filter}(b,a,\text{gripTx10}); \]
\[ \text{startPt} = [11000 \ 19000 \ 30000 \ 40000 \ 52500 \ 61500 \ 72500 \ 84500 \ 100000]; \]
\[ \text{ACsig} = \text{filterSig10}( :,4); \]
\[ \text{DCsig} = \text{filterSig10}( :,2); \]
\[ [\text{grip}(10).\text{AC},\text{grip}(10).\text{DC}] = \text{NormaliseSignal} (\text{ACsig}, \text{DCsig}, \text{startPt}); \]

\%
Calculate the mean and standard deviation for AC and DC signal for \( y = 1: \text{length(startPt)} \)
\[ \text{meanAC}(y) = \text{mean}([\text{grip}(1).\text{AC}(y) \ \text{grip}(2).\text{AC}(y) \ \text{grip}(3).\text{AC}(y) ... \ \text{grip}(4).\text{AC}(y) \ \text{grip}(5).\text{AC}(y) \ \text{grip}(6).\text{AC}(y) \ \text{grip}(7).\text{AC}(y) ... \ \text{grip}(8).\text{AC}(y) \ \text{grip}(9).\text{AC}(y) \ \text{grip}(10).\text{AC}(y)]); \]
\[ \text{stdevAC}(y) = \text{std}([\text{grip}(1).\text{AC}(y) \ \text{grip}(2).\text{AC}(y) \ \text{grip}(3).\text{AC}(y) ... \ \text{grip}(4).\text{AC}(y) \ \text{grip}(5).\text{AC}(y) \ \text{grip}(6).\text{AC}(y) \ \text{grip}(7).\text{AC}(y) ... \ \text{grip}(8).\text{AC}(y) \ \text{grip}(9).\text{AC}(y) \ \text{grip}(10).\text{AC}(y)]); \]
\[ \text{meanDC}(y) = \text{mean}([\text{grip}(1).\text{DC}(y) \ \text{grip}(2).\text{DC}(y) \ \text{grip}(3).\text{DC}(y) ... \ \text{grip}(4).\text{DC}(y) \ \text{grip}(5).\text{DC}(y) \ \text{grip}(6).\text{DC}(y) \ \text{grip}(7).\text{DC}(y) ... \ \text{grip}(8).\text{DC}(y) \ \text{grip}(9).\text{DC}(y) \ \text{grip}(10).\text{DC}(y)]); \]
\[ \text{stdevDC}(y) = \text{std}([\text{grip}(1).\text{DC}(y) \ \text{grip}(2).\text{DC}(y) \ \text{grip}(3).\text{DC}(y) ... \ \text{grip}(4).\text{DC}(y) \ \text{grip}(5).\text{DC}(y) \ \text{grip}(6).\text{DC}(y) \ \text{grip}(7).\text{DC}(y) ... \ \text{grip}(8).\text{DC}(y) \ \text{grip}(9).\text{DC}(y) \ \text{grip}(10).\text{DC}(y)]); \]
grip(4).DC(y) grip(5).DC(y) grip(6).DC(y) grip(7).DC(y) ...
  grip(8).DC(y) grip(9).DC(y) grip(10).DC(y));
end

end

% Plot errorbar graph of AC & DC
x=-20:5:20; % Identify the range of offset positions

figure
% Plot DC
subplot (2,1,1)
hold on
plot(x,meanDC,'*--')
for i = 1:9
  errorbar(x(i),meanDC(i),stdevDC(i));
end
title('DC signal')
xlabel('Light Source Position (mm)')
ylabel('Relative Amplitude (a.u)')

% Plot AC
subplot (2,1,2)
hold on
plot(x,meanAC,'*--')
for i = 1:9
  errorbar(x(i),meanAC(i),stdevAC(i));
end
title('AC Signal')
xlabel('Light Source Position (mm)')
ylabel('Relative Amplitude (a.u)')

% Plot the AC PPG signals at each position position for filterSig1 dataset
figure
subplot (5,2,1)
plot((1:1000)/100,-filterSig1(5001:6000,4))
title('Left 20mm')

subplot (5,2,2)
plot((1:1000)/100,-filterSig1(16001:17000,4))
title('Left 15mm')

subplot (5,2,3)
plot((1:1000)/100,-filterSig1(26001:27000,4))
title('Left 10mm')

subplot (5,2,4)
plot((1:1000)/100,-filterSig1(37001:38000,4))
title('Left 5mm')

subplot (5,2,5)
plot((1:1000)/100,-filterSig1(45001:46000,4))
title('Origin')

subplot (5,2,6)
plot((1:1000)/100,-filterSig1(53001:54000,4))
title('Right 5mm')
title('Right 10mm Offset')
[pos8,lag8]=xcorr(filterSig1(75000:76000,4),'coeff');
subplot(5,1,4)
plot(lag8/100,pos8)
axis([0 10 -1 1])
title('Right 15mm Offset')

[pos9,lag9]=xcorr(filterSig1(87000:88000,4),'coeff');
subplot(5,1,5)
plot(lag9/100,pos9)
axis([0 10 -1 1])
title('Right 20mm Offset')
xlabel('Lag time (s)')
Function $\text{NormaliseSignal}$

1. **Input**: AC signal, DC signal, Start points
2. Identify the number of start points
3. Identify 30% of data for each start point
4. Calculate Mean DC signal
5. Calculate AC/DC for AC PPG
6. Envelope Detector
7. Moving Average
8. Initialise starting positions
9. Completed all starting positions?
10. Yes
    - Output: Normalise Mean envelope peak-to-peak, Normalise Mean DC signal
    - STOP
11. No
    - Calculate the Normalise Mean envelope peak-to-peak
    - Calculate the Normalise Mean DC signal
    - Initialise starting positions
    - Completed all positions?
    - Yes
    - No
function [normAC,normDC,meanDC,pk2pkAC] = NormaliseSignal(AC,DC,startPt)

% Compute the number of start points
dataSize = length(startPt);

% Compute the mean amplitude for AC and DC for each offset positions
for i = 1:dataSize
    sigRange = startPt(i):(startPt(i)+3000); % Identify 30s of signal
    meanDC(i) = mean(DC(sigRange)); % Calculate mean DC
    ACDC = AC(sigRange)./meanDC(i); % Calculate AC/DC
    [posEnv,negEnv] = EnvelopeDetector(ACDC); % Detect envelope for AC signal
    pk2pkAC(i) = MovingAverage(posEnv,negEnv); % Calculate AC peak-to-peak
end

% Compute the normalised mean amplitude for each offset positions
for x = 1:dataSize
    normAC(x) = pk2pkAC(x)/pk2pkAC(5); % Compute normalised AC
    normDC(x) = meanDC(x)/meanDC(5); % Compute normalised DC
end
Function *EnvelopeDetector*

START

Input: AC data, Rate of fall-off, Maximum of change, Initial value, Exponential constant

Exponential rise specified? Yes
- Set Exponential constant = 2

No

Initial start specified? Yes
- Set Initial value = 50

No

Maximum rate of change specified? Yes
- Set to 0.5% of minimum signal

No

Rate of fall-off specified? Yes
- Set to 5e-5 of maximum signal

No

- Initialise variables, counters and flags

- Compute maximum and minimum signal

End of signal length? Yes
- Set tracking positive envelope flag

No

Current signal > Previous positive envelope? Yes
- Calculate adaptive threshold

No

End of signal: length? No

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Output: Positive envelope, Negative envelope

STOP
function [EnvelopePos, EnvelopeNeg] = EnvelopeDetector(PPGsig, Tau, MaxRate, Start, ExpRise)

% Find exponential rise
if nargin < 5
    ExpRise = 2;
end

% Find initial cond in first 50 samples - spike sensitive
if nargin < 4
    Start = 50;
end

% Find maximum allowed rate of change - spike suppression
if nargin < 3
    MaxRate = 0.005*max(PPGsig); % auto normalize
end

% Find rate of fall off
if nargin < 2
    Tau = 5e-5*max(PPGsig);
end

% Initialise variables, counters and flags
flagTrackPos=0; % track positive flag
flagTrackNeg=0; % track negative flag
lastTrackedPos=1; % last 'tracked' positive point
lastTrackedNeg=1; % last 'tracked' negative point
flagLimitPos=0; % positive limit flag
flagLimitNeg=0; % negative limit flag
counterMaxRateNeg=1; % negative counter for max rate
counterMaxRatePos=1; % positive counter for max rate

% Identify maximum and minimum signal
EnvelopePos(1)= max(PPGsig(1:Start));
EnvelopeNeg(1)= min(PPGsig(1:Start));

for t=2:length(PPGsig)
    % Check if current sample point > previous sample of positive envelope
    if PPGsig(t)>EnvelopePos(t-1)
        flagTrackPos=1; % Set tracking positive flag
        % Calculate adaptive threshold
        AdaptiveThresh = EnvelopePos(lastTrackedPos)+MaxRate*counterMaxRatePos;
        % Check current point is greater than adaptive threshold (spike suppression)
        if PPGsig(t) <= AdaptiveThresh
            EnvelopePos(t) = PPGsig(t); % 'Follow' PPG signal
            flagLimitPos = 0; % Clear limit positive flag
        else
            EnvelopePos(t) = AdaptiveThresh; % Limit envelope to adaptive threshold
        end
    end
end

% Calculate adaptive threshold
AdaptiveThresh = EnvelopePos(lastTrackedPos)+MaxRate*counterMaxRatePos;
% 'Follow' PPG signal
flagLimitPos = 0; % Clear limit positive flag
else
    EnvelopePos(t) = AdaptiveThresh; % Limit envelope to adaptive threshold
end
flagLimitPos = 1; % Set limit positive flag
end
lastTrackedPos = t; % Record 'tracked point'
else
EnvelopPos(t) = EnvelopPos(t-1) - Tau*(t-lastTrackedPos); % Fall algorithm
if flagTrackPos == 1
flagTrackPos = 0; % Clear tracking positive flag
if flagLimitPos == 1
    counterMaxRatePos = counterMaxRatePos + ExpRise; % Compute positive max rate of change
else
    counterMaxRatePos = 1; % Set maximum rate of change counter
end
end

% Check if current sample point < previous sample of negative envelope
if PPGsig(t) < EnvelopeNeg(t-1)
flagTrackNeg = 1; % Set tracking negative flag
% Calculate adaptive threshold
AdaptiveThresh = EnvelopeNeg(lastTrackedNeg) - MaxRate * counterMaxRateNeg;
% Check current point is less than adaptive threshold (spike suppression)
if PPGsig(t) <= AdaptiveThresh
    EnvelopeNeg(t) = PPGsig(t); % 'Follow' PPG signal
    flagLimitNeg = 0; % Clear limit negative flag
else
    EnvelopeNeg(t) = AdaptiveThresh; % Limit envelope to adaptive threshold
    flagLimitNeg = 1; % Set limit negative flag
end
lastTrackedNeg = t; % Record 'tracked point'
else
EnvelopeNeg(t) = EnvelopeNeg(t-1) + Tau*(t-lastTrackedNeg); % Fall algorithm
if flagTrackNeg == 1
flagTrackNeg = 0; % Clear tracking negative flag
if flagLimitNeg == 1
    counterMaxRateNeg = counterMaxRateNeg + ExpRise; % Compute negative max rate of change
else
    counterMaxRateNeg = 1; % Set maximum rate of change counter
end
end
end
Function MovingAverage

START

Input Positive envelope, Negative envelope

Compute length of envelope

Set moving average sample

Calculate Moving average for positive envelope

End of data length?

Yes

Calculate value for Peak-to-peak envelope

End of data length?

Yes

Calculate value for Mean envelope peak-to-peak

Output Mean envelope peak-to-peak

STOP

No

End of data length?

Yes

Calculate Moving average for negative envelope

No

End of data length?
Main Program AnalyseCompareAC

START

Extract Signal?

Completed all data sets?

Yes

Compute number of data sets

Set sampling frequency

Initialise start of data sets

End of signal length?

Yes

Peak position for contact signal between start & end point?

Yes

Store contact peak position

Increment contact counter

No

Peak position for non-contact signal between start & end point?

Yes

Store non-contact peak position

Increment non-contact counter

No

No

End of signal length?

No

End of data sets?

No

End data set?

Yes

End of data set?

Yes

End of data set?

Yes

End of data set?

Yes

End of data set?

Yes

End of data set?
% Filename: AnalyseCompareAC  
% A program to identify, analyse and compare the position of the peak in the AC signal  
% between contact and non-contact PPG  
% This program is repeated for non-contact transmission and reflection PPG system.  

% Extract filtered PPG signal  
load TxAC01  
[rem(01).AC,con(01).AC,rem(01).DC,con(01).DC] = ExtractSignal(TxAC01);  
load TxAC02  
[rem(02).AC,con(02).AC,rem(02).DC,con(02).DC] = ExtractSignal(TxAC02);  
load TxAC03  
[rem(03).AC,con(03).AC,rem(03).DC,con(03).DC] = ExtractSignal(TxAC03);  
load TxAC04  
[rem(04).AC,con(04).AC,rem(04).DC,con(04).DC] = ExtractSignal(TxAC04);  
load TxAC05  
[rem(05).AC,con(05).AC,rem(05).DC,con(05).DC] = ExtractSignal(TxAC05);  
load TxAC06  
[rem(06).AC,con(06).AC,rem(06).DC,con(06).DC] = ExtractSignal(TxAC06);  
load TxAC07  
[rem(07).AC,con(07).AC,rem(07).DC,con(07).DC] = ExtractSignal(TxAC07);  
load TxAC08  
[rem(08).AC,con(08).AC,rem(08).DC,con(08).DC] = ExtractSignal(TxAC08);  
load TxAC09  
[rem(09).AC,con(09).AC,rem(09).DC,con(09).DC] = ExtractSignal(TxAC09);  
load TxAC10  
[rem(10).AC,con(10).AC,rem(10).DC,con(10).DC] = ExtractSignal(TxAC10);  
load TxAC11  
[rem(11).AC,con(11).AC,rem(11).DC,con(11).DC] = ExtractSignal(TxAC11);  
load TxAC12  
[rem(12).AC,con(12).AC,rem(12).DC,con(12).DC] = ExtractSignal(TxAC12);  
load TxAC13  
[rem(13).AC,con(13).AC,rem(13).DC,con(13).DC] = ExtractSignal(TxAC13);  
load TxAC14  
[rem(14).AC,con(14).AC,rem(14).DC,con(14).DC] = ExtractSignal(TxAC14);  

% Compute number of data sets  
sets = length(rem);  
% Set sampling frequency  
freq = 100;  

% Identify peak positions in the AC signal  
for i = 1:sets  
    [PeakPos(i).con] = FindPeak(con(i).AC);  
    [PeakPos(i).rem] = FindPeak(rem(i).AC);  
end  

% Locate peak points within lower and upper limit  
for i = 1:sets  
    startPt = 50*freq;  
    endPt = 110*freq;  
    PkThreshold = 125;  
    a = 1;
b = 1;
c = 1;
conFlag = 2;
remFlag = 2;

% Identify minimum size data between PeakPos of contact and non-contact
numPeak = min(length(PeakPos(i).con), length(PeakPos(i).rem));

% Identify peak position between the start and end point (60 seconds of data)
for j = 1:numPeak
    if (PeakPos(i).con(j)>startPt) & (PeakPos(i).con(j)<endPt) % Find contact's peak
        PeakPosRange(i).con(a) = PeakPos(i).con(j);
        a = a+1;
    end
    if (PeakPos(i).rem(j)>startPt) & (PeakPos(i).rem(j)<endPt) % Find non-contact's peak
        PeakPosRange(i).rem(b) = PeakPos(i).rem(j);
        b = b+1;
    end
end

% Find the minimum size of data between contact and non-contact
minSize(i) = min(length(PeakPosRange(i).con), length(PeakPosRange(i).rem));

for m = 1:minSize(i)
    if (remFlag < minSize(i)) & (conFlag < minSize(i))
        % Calculate the difference and mean between contact and non-contact signal
        if (PeakPosRange(i).rem(remFlag)+PeakPosRange(i).rem(remFlag-1)<PkThreshold)&...  
            (PeakPosRange(i).con(conFlag)-PeakPosRange(i).con(conFlag-1)<PkThreshold)
            diffPeak(i).Pk(c) = PeakPosRange(i).rem(remFlag)/freq -...
            PeakPosRange(i).con(conFlag)/freq;
            meanPeak(i).pos(c) = mean([PeakPosRange(i).con(conFlag)/freq ...
                PeakPosRange(i).rem(remFlag)/freq]);
            conFlag = conFlag+1;
            remFlag = remFlag+1;
            c = c+1;
        end
    end

% Calculate mean of the difference in peak position
meanDiff(i) = mean(diffPeak(i).Pk);
% Calculate standard deviation of the difference in peak position
stdDiff(i) = std(diffPeak(i).Pk);
% Calculate standard error, upper and lower limit
stdError(i) = sqrt((stdDiff(i)^2)/length(diffPeak(i).Pk));
upLimitPeak(i) = meanDiff(i)+2*stdDiff(i);
lowLimitPeak(i) = meanDiff(i)-2*stdDiff(i);
end

% Mean Difference Across all Subjects (for thesis!)
meanPos = mean(meanDiff);
stdPos = std(meanDiff);
stdErr = sqrt((stdPos^2)/length(meanDiff));
upLimitPos = meanPos+2*stdPos;
lowLimitPos = meanPos-2*stdPos;

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% Plot 5 seconds comparison plot for PPG signal from contact and non-contact PPG
figure
hold on
plot((1:500)/100,con(1).AC(501:1000),"k")
plot((1:500)/100,rem(1).AC(501:1000),"k--")

% Plot the Bland Altman for amplitude position measured by contact and non-contact PPG
figure
hold on
plot(meanPeak(1).pos,(diffPeak(1).Pk*1000),"s")
plot(meanPeak(1).pos,(meanDiff(1)*1000))
plot(meanPeak(1).pos,(upLimitPeak(1)*1000),meanPeak(1).pos,(lowLimitPeak(1)*1000))

% Plot the Bland Altman for all data sets for amplitude position measured by contact and non-contact PPG
figure
hold on
plot(1:sets,(meanDiff*1000),"s")
plot(1:sets,(meanPos*1000))
plot(1:sets,(upLimitPos*1000),1:sets,(lowLimitPos*1000))
Function ExtractSignal

START

Input: PPG signal, Signal range

Compute coefficients for 10th order low pass Butterworth filter

Apply filter coefficients on PPG signal

Separate the AC and DC component of contact and non-contact from the PPG signal

Signal range provided?

Yes

Compute AC/DC of the Signal range for AC of contact and non-contact signal

MoveAvgDC

Output: Contact AC, Contact DC, Non-contact AC, Non-Contact DC

STOP

No

Compute AC/DC of the entire PPG signal for AC of contact and non-contact signal

MoveAvgDC

Output: Contact AC, Contact DC, Non-contact AC, Non-Contact DC

STOP
% Filename: ExtractSignal
% A function to extract and format AC and DC signal
% Input: PPG signal, Signal range
% Output: Non-contact AC, Contact AC, Non-contact DC, Contact DC

function [remAC,conAC,remDC,conDC] = ExtractSignal(PPGsig,SigRange);

% Compute coefficient of Butterworth filter
[b,a] = butter(10,0.1);
filtSig = filter(b,a,PPGsig);

% Identify AC and DC components from PPG signal
con_AC = filtSig(:,5);
con_DC = PPGsig(:,3);
rem_AC = filtSig(:,4);
rem_DC = PPGsig(:,2);

if nargin > 1
    % Process AC and DC signal for the specified signal range
    conAC = (-con_AC(SigRange))./con_DC(SigRange);
    remAC = (-rem_AC(SigRange))./rem_DC(SigRange);
    conDC = MoveAvgDC(con_DC(SigRange));
    remDC = MoveAvgDC(rem_DC(SigRange));
else
    % Process AC and DC signal for the entire signal range
    conAC = (-con_AC)./con_DC;
    remAC = (-rem_AC)./rem_DC;
    conDC = MoveAvgDC(con_DC);
    remDC = MoveAvgDC(rem_DC);
end
Function *MoveAvgDC*

**START**

Input: DC signal

Compute length of DC signal

Initialise start of data

End of data length?

No

Compute Moving average data

Output: Average DC

Yes

STOP
function [MeanDCsig] = MoveAvgDC(DCsig)

datalength = length(DCsig); % Compute the length of the signal
sample = 100; % Number of samples averaged over

% Find moving average for the data
for x = 1:(datalength-sample)
    MeanDCsig(x) = mean(DCsig(x:(x+sample)));
end
Function **FindPeak**

1. **START**
2. Input: AC signal, Threshold
3. Compute Threshold = mean absolute AC signal
   - No: **Threshold specified?**
   - Yes: Compute Approximate derivative for AC signal
4. Compute Length of derivative signal
5. Initialise Minimum peak distance and Amplitude index
6. Initialise start of data
7. End of data length?
8. Output: Position of peak amplitude
9. **STOP**
function [PeakPosition] = FindPeak(ACsig, threshold);

if nargin < 2 % Check for threshold
    threshold = mean(abs(ACsig));
end

% Compute approximate derivative
derivativeAC = diff(ACsig);
% Compute length of derivative signal
dataLength = length(derivativeAC);

% Initialise variables
MinDistance = 55; % Minimum distance between peak amplitude
index = 1; % Index for amplitude position

for i = 2 : dataLength - 1
    % Detect peak by finding the negative slope zero crossing of the derivative ACsignal
    if (derivativeAC(i-1) > 0) & (derivativeAC(i+1) < 0) & (ACsig(i) > threshold)
        if index == 1
            PeakPosition(index) = i + 1; % Position of peak
            index = index + 1; % Increment index
        else
            % Ensure that peak is not noise from signal trough
            if i - PeakPosition(index-1) > MinDistance
                PeakPosition(index) = i + 1;
                index = index + 1;
            end
        end
    end
end
Main Program AnalyseVOP

START

Extract Signal

Completed all data sets?

Compute number of data sets

Set sampling frequency

Initialize start of data sets

End of data sets?

Yes

No

Normalize the VOP curve for contact and non-contact DC signal

Plot venous occlusion curve for contact and non-contact curve

Initialize start of data sets

End of data sets?

Yes

No

Calculate tangent of arterial inflow

End of data sets?

Yes

No

Calculate mean and standard deviation for the DC signal

Calculate the mean and difference of the mean DC signal

Calculate mean, standard error and limit of agreement of mean DC signal

Bland Altman plot for mean DC signal

Identify start point for arterial inflow

Initialize start of data sets

End of data sets?

Yes

No

Calculate mean and difference of arterial inflow between contact and non-contact signal

Calculate mean, standard error and limit of agreement of arterial inflow

Bland Altman plot for arterial inflow

Identify start and end point to calculate arterial inflow

Initialize start of data sets

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End of data sets?

No

Identify minimum DC amplitude for contact and non-contact signal

Yes

Initiate start of data set

End of data sets?

No

Calculate maximum change of DC amplitude and its mean difference between contact and non-contact

Yes

Calculate mean, standard error and limit of agreement of venous capacitance

Bland Altman plot for venous capacitance

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% Filename: AnalyseCompareVOP
% A program to identify, analyse and compare the VOP curve in the DC signal
% between contact and non-contact PPG
% This program is repeated for non-contact transmission and reflection PPG system.

% Extract filtered PPG signal
load TxVOP01 % Load Matlab file
[rem(01).AC,con(01).AC,rem(01).DC,con(01).DC] = ExtractSignal(TxVOP01);
load TxVOP02
[rem(02).AC,con(02).AC,rem(02).DC,con(02).DC] = ExtractSignal(TxVOP02);
load TxVOP03
[rem(03).AC,con(03).AC,rem(03).DC,con(03).DC] = ExtractSignal(TxVOP03);
load TxVOP04
[rem(04).AC,con(04).AC,rem(04).DC,con(04).DC] = ExtractSignal(TxVOP04);
load TxVOP05
[rem(05).AC,con(05).AC,rem(05).DC,con(05).DC] = ExtractSignal(TxVOP05);
load TxVOP06
[rem(06).AC,con(06).AC,rem(06).DC,con(06).DC] = ExtractSignal(TxVOP06);
load TxVOP07
[rem(07).AC,con(07).AC,rem(07).DC,con(07).DC] = ExtractSignal(TxVOP07);
load TxVOP08
[rem(08).AC,con(08).AC,rem(08).DC,con(08).DC] = ExtractSignal(TxVOP08);
load TxVOP09
[rem(09).AC,con(09).AC,rem(09).DC,con(09).DC] = ExtractSignal(TxVOP09);
load TxVOP10
[rem(10).AC,con(10).AC,rem(10).DC,con(10).DC] = ExtractSignal(TxVOP10);
load TxVOP11
[rem(11).AC,con(11).AC,rem(11).DC,con(11).DC] = ExtractSignal(TxVOP11);
load TxVOP12
[rem(12).AC,con(12).AC,rem(12).DC,con(12).DC] = ExtractSignal(TxVOP12);
load TxVOP13
[rem(13).AC,con(13).AC,rem(13).DC,con(13).DC] = ExtractSignal(TxVOP13);
load TxVOP14
[rem(14).AC,con(14).AC,rem(14).DC,con(14).DC] = ExtractSignal(TxVOP14);

% Compute number of data sets
sets = length(rem);
% Set sampling frequency
freq = 100;

% Normalise the DC signal to the base of the curve
for i = 1:sets
    DC(i).con = con(i).DC./con(i).DC(1);
    DC(i).rem = rem(i).DC./rem(i).DC(1);
end

% Plot comparison venous occlusion curve
figure
hold on
plot((1:2:1000)/freq,DC(1).con,'k')
plot((1:2:1000)/freq,DC(1).rem,'k--')
title('Venous Occlusion Curve')
xlabel('Time (s)')
ylabel('Relative Amplitude (a.u)')
% Calculate the mean and standard deviation for both contact and non-contact signal for i = 1:sets
    % Find the DC mean of contact and non-contact curve
    meanConDC(i) = mean(DC(i).con);
    meanRemDC(i) = mean(DC(i).rem);
    % Find the standard deviation of contact and non-contact curve;
    stdConDC(i) = std(DC(i).con);
    stdRemDC(i) = std(DC(i).rem);
    % Calculate mean curve of contact and non-contact PPG signal
    mmMeanDC(i) = mean([meanRemDC(i) meanConDC(i)]);
    % Calculate difference of mean curve between contact and non-contact
    diffMeanDC(i) = meanRemDC(i)-meanConDC(i);
end
% Calculate parameters mean DC Bland Altman plot
meanDiffDC = mean(diffMeanDC);
stdMeanDiff = std(diffMeanDC);
upLimitMeanDC = meanDiffDC+1.96*stdMeanDiff;
lowLimitMeanDC = meanDiffDC-1.96*stdMeanDiff;
% Bland Altman plot for mean DC
figure
hold on
plot(mmMeanDC,diffMeanDC,'ks')
plot(mmMeanDC,upLimitMeanDC)
plot(mmMeanDC,lowLimitMeanDC)
plot(mmMeanDC,meanDiffDC)
title ('Mean DC')
xlabel ('Average between Non-Contact and Contact PPG (a.u.)')
ylabel ('Difference between Non-Contact and Contact PPG (a.u.)')

% Calculate arterial inflow for contact & non-contact VOP curve
startPtAl = 60*freq; % Identify start point for calculating arterial inflow
% Identify the start and end point for calculating arterial inflow
for i = 1:sets
    remAlst(i) = DC(i).rem(60*freq);
    remAlend(i) = DC(i).rem(63*freq);
    conAlst(i) = DC(i).con(60*freq);
    conAlend(i) = DC(i).con(63*freq);
end
% Calculate the tangent in the first 3 seconds after occlusion
RemAI = (abs(remAlend-remAlst)/30)*60;
ConAI = (abs(conAlend-conAlst)/30)*60;
for i = 1:sets
    % Calculate mean between non-contact and contact arterial inflow
    meanAI(i) = mean([RemAI(i) ConAI(i)]);
    % Calculate difference between non-contact and contact arterial inflow
    diffAI(i) = RemAI(i) - ConAI(i);
end
% Calculate parameters for arterial inflow Bland Altman plot
meanDiffAI = mean(diffAI);
stdDiffAI = std(diffAI);
upLimitAI = meanDiffAI + 1.96*stdDiffAI;
lowLimitAI = meanDiffAI - 1.96*stdDiffAI;
% Bland Altman plot for arterial inflow
figure
hold on
plot(meanAI,diffAI,'s')
plot(meanAI,meanDiffAI)
plot(meanAI,upLimitAI)
plot(meanAI,lowLimitAI)
title ('Arterial Inflow')
xlabel ('Average between Non-Contact and Contact PPG (%)/min)'
ylabel ('Difference between Non-Contact and Contact PPG (%)/min)

% Calculate venous capacitance for contact and non-contact VOP curve
% Calculate minimum DC amplitude for contact and non-contact
for i = 1:sets
    minPtemVC = find(DC(i).rem == min(DC(i).rem));
    minPconVC = find(DC(i).con == min(DC(i).con));
end
% Calculate the maximum change in DC measurement
for i = 1:sets
    remVC(i) = DC(i).rem(minPtemVC(i)) - DC(i).rem(startPtAl(i));
    conVC(i) = DC(i).con(minPconVC(i)) - DC(i).con(startPtAl(i));
% Calculate mean and difference of venous capacitance measurement
meanVC(i) = mean([remVC(i) conVC(i)]);
diffVC(i) = remVC(i)-conVC(i);
end
% Calculate parameters for venous capacitance Bland Altman plot
meanDiffVC = mean(diffVC);
stdDiffVC = std(diffVC);
upLimitVC = meanDiffVC + 1.96*stdDiffVC;
lowLimitVC = meanDiffVC - 1.96*stdDiffVC;
% Bland Altman plot for venous capacitance
figure
hold on
plot(meanVC,diffVC,'s')
plot(meanVC,meanDiffVC)
plot(meanVC,upLimitVC)
plot(meanVC,lowLimitVC)
title ('Venous Capacitance')
xlabel ('Average between Non-Contact and Contact PPG (%)')
ylabel ('Difference between Non-Contact and Contact PPG (%)')

% Calculate venous outflow for contact and non-contact VOP curve
stPVO = 180*freq;
% Calculate the tangent in the first second after deflation
for i = 1:sets
    remVO(i) = (abs(DC(i).rem(180*freq)-DC(i).rem(181*freq)))*60;
    conVO(i) = (abs(DC(i).con(180*freq)-DC(i).con(181*freq)))*60;
% Calculate the mean and difference in venous outflow
meanVO(i) = mean([remVO(i) conVO(i)]);
diffVO(i) = remVO(i)-conVO(i);
end
% Calculate parameters for venous outflow Bland Altman plot
meanDiffVO = mean(diffVO);
stdDiffVO = std(diffVO);
upLimitVO = meanDiffVO + 1.96*stdDiffVO;
lowLimitVO = meanDiffVO - 1.96*stdDiffVO;

% Bland Altman plot for venous outflow
figure
hold on
plot(meanVO,diffVO,'s')
plot(meanVO,meanDiffVO)
plot(meanVO,upLimitVO)
plot(meanVO,lowLimitVO)
title('Venous Outflow')
xlabel('Average between Non-Contact and Contact PPG (%/min)')
ylabel('Difference between Non-Contact and Contact PPG (%/min)')
APPENDIX J

Autocorrelation Plots for Indirect Exposure to Direct Coupling using Non-Contact Transmission Photoplethysmography
**Data Set 1:** This data set shows that arterial PPG signals were successfully recorded at the left offset positions of 15mm and 20mm, as well as at the right offset position of 5mm, 10mm and 15mm.
Data Set 2: It can be observed from the data set that PPG AC signals was successfully recorded from five positions, on left offset at 5mm, 10mm and 15mm, and on the right offset at 5mm and 20mm.
Data Set 3: Besides the left offset position of 20mm, all other positions successfully recorded good signal-to-noise ratio of PPG AC signals.
Data Set 4: The arterial PPG signals were recorded only at three positions in this data set, at the left offset of 15mm, and the right offset of 5mm and 10mm.

![Graph showing arterial PPG signals at different offsets](image-url)