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Estimating pulse wave velocity using mobile phone sensors

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Estimating Pulse Wave Velocity using

Mobile Phone Sensors

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ABSTRACT

Pulse wave velocity has been recognised as an important physiological phenomenon in the human body, and its measurement can aid in the diagnosis and treatment of chronic diseases. It is the gold standard for arterial stiffness measurements, and it also shares a positive relationship with blood pressure and heart rate. There exist several methods and devices via which it can be measured. However, commercially available devices are more geared towards working health professionals and hospital settings, requiring a significant monetary investment and specialised training to operate correctly. Furthermore, most of these devices are not portable and thus generally not feasible for private home use by the common individual. Given its usefulness as an indicator of certain physiological functions, it is expected that having a more portable, affordable, and simple to use solution would present many benefits to both end users and healthcare professionals alike.

This study investigated and developed a working model for a new approach to pulse wave velocity measurement, based on existing methods, but making use of novel equipment. The proposed approach made use of a mobile phone video camera and audio input in conjunction with a Doppler ultrasound probe. The underlying principle is that of a two-point measurement system utilising photoplethysmography and electrocardiogram signals, an existing method commonly found in many studies. Data was collected using the mobile phone sensors and processed and analysed on a computer. A custom program was developed in MATLAB that computed pulse wave velocity given the audio and video signals and a measurement of the distance between the two data acquisition sites. Results were compared to the findings of previous studies in the field, and showed similar trends.

As the power of mobile smartphones grows, there exists potential for the work and methods presented here to be fully developed into a standalone mobile application, which would bring forth real benefits of portability and cost-effectiveness to the prospective user base.

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LIST OF DEFINITIONS

The following definitions of terms used in this thesis have been sourced from the Oxford Dictionaries website (http://oxforddictionaries.com):

Ambulatory: relating to or adapted for walking.

Artery: any of the muscular-walled tubes forming part of the circulation system by which blood (mainly that which has been oxygenated) is conveyed from the heart to all parts of the body.

Arteriosclerosis: the thickening and hardening of the walls of the arteries, occurring typically in old age.

Auscultation: the action of listening to sounds from the heart, lungs, or other organs, typically with a stethoscope as a part of medical diagnosis.

Blood vessel: a tubular structure carrying blood through the tissues and organs.

Diastole: the phase of the heartbeat when the heart muscle relaxes and allows the chambers to fill with blood.

Dysrhythmia: abnormality in a physiological rhythm, especially in the activity of the brain or heart.

Electrocardiogram: a record or display of a person's heartbeat produced by electrocardiography.

Hypertension: abnormally high blood pressure.

Non-intrusive: (of medical procedures) not involving the introduction of instruments into the body.

Plethysmograph: an instrument for recording and measuring variation in the volume of a part of the body, especially as caused by changes in blood pressure.

Pulse: a rhythmical throbbing of the arteries as blood is propelled through them, typically as felt in the wrists or neck.

Pulse oximeter: an instrument that measures the proportion of oxygenated haemoglobin in the blood in pulsating vessels, especially the capillaries of the finger or ear.

Sphygmomanometer: an instrument for measuring blood pressure, typically consisting of an inflatable rubber cuff which is applied to the arm and connected to a column of mercury next to a graduated scale, enabling the determination of systolic and diastolic blood pressure by increasing and gradually releasing the pressure in the cuff.

Systole: the phase of the heartbeat when the heat muscle contacts and pumps blood from the chambers into the arteries.

Tonometer: an instrument for measuring the pressure in a part of the body, such as the eyeball or a blood vessel.

Transducer: a device that converts variations in a physical quantity, such as pressure or brightness, into an electrical signal, or vice versa.

Ultrasound: sound or other vibrations having an ultrasonic frequency, particularly as used in medical imaging.

Viscosity: the state of being thick, sticky, and semi-fluid in consistency, due to internal friction.

Waveform: a curve showing the shape of a wave at a given time.

LIST OF ACRONYMS

BMI	Body Mass Index
BP	Blood pressure
ECG	Electrocardiogram
GPRS	General Packet Radio Service
GSM	Global System for Mobile communications
LED	Light Emitting Diode
mm Hg	Millimeters of mercury
PDA	Personal Digital Assisstant
PPG	Photoplethysmography
PTT	Pulse transit time
PWV	Pulse wave velocity
QBIC	Cubic Belt-Integrated Computer
SSL	Secure Socket Layer
TDI	Tissue Doppler Imaging
TIU	Telephone Interface Unit

1.0   INTRODUCTION

"The pulse wave is a complex physiological phenomenon observed and detected in blood circulation" (Korpas, Halek & Dolezal, 2009). Contraction of the heart during the course of a normal heartbeat causes a certain amount of blood to be ejected and introduced into the arteries due to the transformation between kinetic and potential energy of each segment of ejected blood (Korpas et al., 2009). The contour of the pulse wave varies during circulation, affected by physiological factors such as age, height, body mass index (BMI), and heart rate. Various invasive methods exist for the detection of the pulse wave (Korpas et al., 2009), but increased interest and steady growth in the number of studies in the field (Lim & Lip, 2008) has led to the development of non-invasive methods as well.

Pulse wave velocity (PWV) is "a basic parameter in the dynamics of pressure and flow waves travelling in arteries" (Harada, Okada, Niki, Chang & Sugawara, 2002). Simply defined, PWV is a measure of the amount of time taken by the pulse wave to travel a specific distance through the arterial tree (Lim & Lip, 2008). The velocity of the propagation of the pulse wave varies depending on the site of measurement, and is primarily affected by the elasticity or stiffness of the arterial walls. The stiffer the arteries, the faster the PWV, and vice versa (Salvi et al., 2008). As a person continues to age, the arteries lose their elasticity and gradually get stiffer with time. Other pathological factors such as diabetes, hypertension, and arteriosclerosis can further speed up this process, thereby resulting in a much faster PWV in older people (Korpas et al., 2009). Due to this well-established correlation, PVW is taken to be the primary indicator of arterial stiffness (Harada et al., 2002; Kim et al., 2007; Salvi et al., 2008; Lim & Lip, 2008;

[1]

Baguet et al., 2003), and is hence considered the gold standard measurement in this area.

However, the measurement of PWV can provide benefits that extend beyond the sole function of indicating arterial stiffness. Studies have indicated that a positive relationship exists between both PWV and blood pressure (Korpas et al., 2009), as well as PWV and heart rate (Lantelme, Mestre, Lievre, Gressard, & Millon, 2002). A brief description of these vital signs follows:

- Heart rate refers to the number of times the heart beats over a unit period of time, which is governed by the number of contractions of the lower ventricles ("Definition of heart rate", 2011). It is usually expressed in units of beats per minute (bpm). The normal resting heart rate for a healthy adult lies between 60 – 100 bpm (Vorvick, 2011).

- Blood pressure (BP) refers to the pressure exerted by the circulating blood against the blood vessel walls (Wallymahmed, 2008). It is represented by two numbers which refer to the two different phases of a heartbeat; the systolic phase, which is when the heart muscles contract to pump blood, and the diastolic phase, which is when the heart muscles relax and refill with blood. Measurements are taken as systolic pressure over diastolic pressure, represented in units of millimetres of mercury (mm Hg). A BP reading of 120/80 mm Hg is considered the norm for a healthy adult (Wallymahmed, 2008), and any results equal to or higher than 140/90 mm Hg over a period of time indicates hypertension (Tomas, 2003).

Thus, according to the aforementioned relationship, an increase in BP or heart rate should be met with a corresponding rise in PWV, and vice versa. Hence PWV measurement can potentially be used to reveal trends in blood pressure and heart rate, without the need to actually measure these vital signs separately. In cases which simply require detection of change in these vitals rather than their absolute values, there is the obvious benefit of saving time spent on taking multiple measurements with different devices. Thus, PWV might also be used as a measurement index in the detection and treatment of chronic ailments where blood pressure or cardiac output is a major determinant of one's health condition.

1.1   Background

As mentioned earlier, several different methods exist for both invasive and non-invasive measurement of PWV, and each relies on different principles and makes use of different types of equipment. Techniques such as photoplethysmography, Doppler ultrasound, electrocardiogram, and tonometry are the most common, evidenced by their widespread use in many other studies, and these are discussed in more detail in Section 2.2. Dedicated PWV monitors such as the SphygmoCor (AtCor Medical, Australia) and Complior (Alam Medical, France) systems are also available, making use of pressure sensors to compute PWV along a given arterial segment. However, these methods and devices suffer from a major limitation; their usage is restricted to hospitals and clinics, and they cannot lend themselves to everyday, individual use in a common household. This is due to the fact that these methods make use of medical-grade equipment which is bulky, extremely expensive to obtain, and requires professional training

and knowledge to operate in order to get accurate results. Thus, unlike the many digital monitors available for home measurement of BP and heart rate, there is no simple, portable, or affordable solution for PWV measurement.

These facts set the stage for this current research undertaking. Being a gold standard measure of arterial stiffness and a possible predictor of other vital signs, it would be of significant benefit if the ability to measure PWV was brought into the common household. There exists a need for a new type of system that can be aimed at the common individual, one that can be operated without having specialised medical knowledge or spending thousands of dollars. This is the central focus of this study; to investigate and develop a new approach to PWV measurement by using commonly available and relatively inexpensive devices. The proposed approach makes use of a mobile phone and portable Doppler ultrasound sensor to acquire the needed data for PWV computation. The study investigates the feasibility of this approach via development and testing of a prototype model on a computer. This provides a working model and basis for future work in the field, which could potentially result in the development of a fully functional mobile phone application, thereby making it a portable, simple, and cost-effective telemedicine system.

1.2   Purpose

The focus of this study is to investigate the possibility of a new technique for PWV measurement. The proposed approach makes use of mobile phone sensors together with a Doppler ultrasound probe for a two-point estimation of PWV. This technique uses well-known principles derived from existing methods in the

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field, the major difference being that the proposed approach makes use of common, easily obtainable devices as opposed to sophisticated and expensive medical equipment. Through the course of this study, a working prototype for non-invasive estimation of PWV using data acquired from the mobile phone was developed. The main aim was to determine if the chosen methods and instruments were capable of producing usable results. The work carried out in this study provides the groundwork and information required to potentially develop a real-world mobile system for PWV measurement in the future. Such a system would allow for personal home monitoring by end users, without the need to pay for expensive medical equipment or undergo training to operate complicated devices.

1.3 Significance

The approach and methods proposed in this study are significant due to the following:

1) The hardware and devices needed for data acquisition are simple, commonly found, and comparatively less expensive than medical devices. Mobile phones are more readily available than ever, and all of them feature the components required; a camera and microphone input. Portable Doppler ultrasound probes are also inexpensive. Hence as a whole, the required setup is much more cost-effective than the commercial PWV monitors, whose prices run into thousands of dollars.

2) Having a mobile phone as part of the setup means the system would also provide the advantage of being able to relay measurement data to remote locations, such as a hospital or clinic, from the end user's current location. This means that a healthcare professional could have near-instant access to patient data, moments after a recording was performed by the user. This could be a great aid in timely diagnosis and would also save the user from making unnecessary trips to visit their doctor.

3) If all operations are performed by the mobile phone, users will effectively be able to monitor their PWV at any place or time, due to the compact setup and use of portable components. The system would thus not impose restrictions on user mobility.

4) The proposed method of data acquisition for PWV measurement is simple and straightforward, such that users would require little formal or technical training to operate it.

1.4   Research Questions

The main research question central to this study was:

- Can PWV be measured with the aid of video and audio signals acquired using a mobile phone and Doppler ultrasound?

The following sub-questions were also answered as part of the investigation:

- How exactly will the audio and video data required for PWV calculations be acquired from the mobile phone sensors?

- How will the acquired data be processed in order to yield information required for PWV calculation?

- What algorithm shall be used to allow for automatic computation of PWV?

The main objective was to develop and present a new working model for PWV estimation. The basic underlying idea is to use a mobile phone to acquire audio of the user's heartbeat and video of a particular region of interest on the body, and then process these in a way that allows calculation of PWV using established formulas.

1.5   Contributions of this Study

In terms of contributing new knowledge, this study has:

1) Identified a suitable set of devices for a simple, portable, and relatively cost effective PWV measurement system.

2) Demonstrated the possibility and detailed the process of calculating heart rate from audio and video recordings obtained from a mobile phone and Doppler ultrasound probe, verified against a commercial heart rate monitor.

3) Described, in detail, a new method and automatic algorithm for a two-point estimation of PWV using audio and video signals acquired from a mobile phone and Doppler ultrasound probe.

4) Resulted in the development of a working prototype application in MATLAB that implements the new algorithm for automated computation of PWV. This prototype may be a potential starting point for the development of a full-fledged mobile system in the future.

5) Showed that PWV can be successfully estimated using an audio and video signal, without the need for expensive, sophisticated equipment. The results obtained from the various experiments performed show similar trends as those published by other studies, which made use of different devices and methods.

6) Identified areas of improvement and avenues for future research and development of the methods and ideas presented here.

1.6   Structure of the Thesis

This thesis is divided into a total of 7 chapters, and this section briefly summarises the content of each of these.

1) *Chapter 1* introduces the main subject matter of this study, i.e. PWV. It provides a brief background to the study, before outlining its purpose and

significance, and then discussing the research questions that the study aims to answer. It also lists the new contributions to knowledge made by the study.

2) *Chapter 2* presents an extensive literature review that covers all aspects of knowledge deemed relevant to this study. It examines a variety of methods and devices used to measure PWV, along with providing a detailed insight into the supplementary topics of BP and mobile health as related to the context of this study.

3) *Chapter 3* discusses the chosen research methodology and lists details of all the materials and instruments used in the study. It also covers details of the new approach for PWV measurement that this study proposes.

4) *Chapter 4* describes the data acquisition and processing methods for the audio and video signals required for PWV computation. The set of experiments performed to rule out certain variables in the acquisition process are also documented here, along with their results.

5) *Chapter 5* presents details on the main focus of this study – PWV measurement using the new proposed approach. This chapter details the steps of the algorithm used in the development of the prototype program in MATLAB for PWV estimation. It also discusses issues encountered during this process, and how these were subsequently resolved. Finally, it documents all the experiments carried out on two subjects to measure their PWV under different conditions, and discusses the results obtained with reference to PWV results from other studies in the field.

6) *Chapter 6* follows up by providing a discussion on the results obtained from the PWV experiments in Chapter 5, as well as outlining avenues for future research based on the work conducted and presented in this study.

7) *Chapter 7* brings the study and this thesis to a close by providing concluding remarks on the work done, and by providing a summary of answers to the research questions and sub-questions that were posed at the beginning in Chapter 1.

1.7   Summary

This chapter has set the basic premise for the study by providing relevant background information and discussing what the study aims to achieve and how this may be of significance in the field of PWV measurement. The questions that need to be answered through this investigation have also been presented here.
In order to work towards providing the required answers, it is necessary to gain some understanding of other work carried out in areas related to the study. To this end, a detailed literature review has been conducted, the findings of which are presented in the next chapter.

2.0 LITERATURE REVIEW

This chapter focuses on providing an insight into existing knowledge in the areas related to this study. BP measurement techniques and devices, various aspects of PWV measurement methods and equations, and some noteworthy work in the field of telemedicine are all covered here. The study draws upon some of the specific knowledge presented in this section.

2.1 Blood Pressure Monitoring: Device Types and Techniques

"The measurement of blood pressure is one of the most common examinations undertaken in family practice and has important health and management consequences for the patient" (Grace, Sabin & Dawes, 2008). There are several guidelines to be followed when taking a BP measurement, some specific to the types of devices being used (e.g., choosing the right cuff size for cuff-based devices) and some generic ones (e.g., five minute rest period prior to measurement, proper placement of patient arm) in order to secure accurate results (Grace et al., 2008). In this section, the different types of BP monitoring devices available as well as some of the most common measurement techniques employed by them shall be examined.

1) *The Korotkoff Method* is one of the earliest techniques developed to measure BP. It was named after Nikolai Korotkoff, a Russian surgeon who first described the technique in 1905. Still widely used today, this method employs a sphygmomanometer, a stethoscope, and an inflatable cuff attached to a

rubber bulb. The steps employed in this method have been outlined in Pickering et al. (2005) as follows:

- The cuff is placed around the patient's arm over the brachial artery and inflated to a point above systolic BP; approximately 30 mm Hg higher (Jilek & Fukushima, 2005). This restricts the blood flow in the arteries temporarily.

- The cuff is then gradually deflated at a rate of about 3 to 5 mm Hg (Jilek & Fukushima, 2005) and the blood starts to flow again. This resumption of blood flow is accompanied by certain sounds which are produced by the combination of turbulent blood flow and arterial wall oscillations, known as Korotkoff sounds. There are different stages of Korotkoff sounds, and these are observed via the stethoscope:

Phase 1 is the appearance of tapping sounds that correspond to the pulse.

Phase 2 is when the sounds get longer and softer.

Phase 3 is when the sounds get crisper and louder.

Phase 4 is when the sounds start to get muffled and softer again.

Phase 5 is marked by the disappearance of the sounds as normal blood flow resumes.

- The person performing the BP monitoring notes the mm Hg reading from the sphygmomanometer at the time of hearing the Phase 1 and Phase 5 Korotkoff sounds, as these phases correspond to the systolic and diastolic BP respectively. This process is known as auscultation.

  Two important drawbacks associated with the Korotkoff technique and the process of auscultation are that one cannot perform it on oneself unaided, and that the technique must be employed by a trained and qualified medical professional to ensure accuracy of results.

2) *Mercury Sphygmomanometers* consist of a glass tube graded in mm Hg, dipped in a bowl of liquid mercury. Also part of the setup is an inflatable cuff connected to a rubber bulb, and a stethoscope. The cuff is wrapped around the patient's arm and then inflated using the bulb, causing the mercury level to rise in the glass tube. The person monitoring the patient places the stethoscope just over the brachial artery in the arm, and performs auscultation to obtain the systolic BP and diastolic BP readings. Due to the fact that these devices are manually calibrated and operated, they remain accurate for several years, and are considered the gold standard against which all other types of BP monitoring devices are tested.

3) *Aneroid Sphygmomanometers* are mechanical devices consisting of metal bellows which expand when the cuff is inflated, and a series of levers that register the pressure readings on a circular dial via a needle (Pickering et al., 2005). They serve as an alternative pressure indicator to the mercury-based sphygmomanometers. The person performing the monitoring makes use of the standard auscultatory method with the aid of a stethoscope to determine

the points at which the Phase 1 and Phase 5 Korotkoff sounds are heard. These devices are inherently less accurate than their mercury counterparts due to their composition of mechanical parts, and hence require frequent recalibration. Their accuracy also tends to suffer if the device is roughly handled or dropped.

With the advent of digital monitoring devices, most of the aforementioned issues can be mitigated. Patients can take their own readings at home without requiring any medical or technical expertise. This is possible because these digital devices calculate BP by measuring oscillations in the arm produced by cuff pressure, and hence no one needs to perform auscultation. This technique and the devices that employ it are discussed below.

1) *The Oscillometric Technique* was developed and first demonstrated in 1876. In this method, oscillations in a BP monitor cuff are recorded during deflation (Pickering et al., 2005) rather than listening for specific sounds, but this also makes it less precise than the auscultatory method. This technique is most commonly used by automated BP monitors. When the artery underlying the cuff is compressed via cuff inflation, it does not emit any pulsations ("Method for the measurement of the blood pressure", 2003). When cuff deflation begins, the pulsations start to occur again, and the pressure recorded at that moment in time is the systolic pressure. As cuff pressure continues to decrease, the oscillations become increasingly significant, and then reach a point where they begin to finally fade. The pressure registered by the monitor at that moment in time refers to the diastolic pressure. The automated devices that employ this technique of BP measurement arrive at the systolic

BP and diastolic BP readings automatically by making use of empirically derived formulas which are programmed into the unit.

2) *Automated Arm Sphygmomanometers* utilise the oscillometric technique to measure BP. They consist of a digital monitoring unit connected to an inflatable cuff worn around the upper arm, which provides an easy reading of systolic BP, diastolic BP and pulse on the electronic display (Nelson et al., 2008). Automated oscillometric instruments make use of various algorithms to measure BP using microprocessors embedded in the unit (Jilek & Fukushima, 2005). Since the readings are digitised, most units come with storage functions that allow a certain number of readings to be saved and retrieved later on. These devices have some limitations in the sense that they are not suitable for people with irregular heartbeats or arterial stiffness, as these conditions hamper measurement accuracy (Nelson et al., 2008). Additionally, the algorithms used to calculate systolic BP and diastolic BP differ by manufacturer and are not generally divulged, so results can vary with brands (Pickering et al., 2005).

3) *Automated Wrist Sphygmomanometers* work on the same principle as automated arm sphygmomanometers but are worn with the cuff around the wrist. They are more compact than arm monitors and offer the advantage of a nearly universal cuff size, as the differences in wrist diameters of people of different sizes and weights is minimal (Pickering et al., 2005). It is, however, worth noting that systolic BP and diastolic BP readings can vary greatly depending on the point above the artery from which they are recorded. These devices are also more susceptible to erroneous measurement due to the inherent difference in the position of the wrist with respect to the heart. The

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devices are typically calibrated to produce a correct reading when held at heart level, but there is "no way of knowing retrospectively whether this was performed when a series of readings are reviewed" (Pickering et al., 2005).

4) *Ultrasound Techniques* make use of methods relying on the Doppler principle to record systolic and diastolic BP via sensors placed on the patient's body (Pickering et al., 2005). The sensors are usually a pair of ultrasound emitters and receivers. These emit sound waves which enter the body through the skin's surface and eventually hit their target and get reflected back to the surface. This reflection or scattering of the sound waves results in a change in frequency in the reflected wave as compared to the original one, and this change may be observed when the sensor picks up the reflected wave (Xu, 2002). This phenomenon is based on what is known as the Doppler Principle or the Doppler Effect, proposed by the Austrian mathematician Christian Johann Doppler in the year 1842. The principle explained how the frequency of waves in a medium underwent an apparent shift if either their source or target were in motion with respect to the medium, and it has given rise to many practical applications since its inception, from meteorological radars to echocardiograms and more (Harris, n.d.). One technique involves the placement of the ultrasound sensors over the brachial artery under the cuff of a sphygmomanometer (Pickering et al., 2005). During cuff deflation, "the movement of the arterial wall at systolic pressure causes a Doppler phase shift in the reflected ultrasound, and diastolic pressure is recorded at the point at which diminution of arterial motion occurs" (Pickering et al., 2005). A variation of the method is used to detect the onset of blood flow, which can provide a measurement of systolic BP in young children and infants (Pickering et al., 2005). Another ultrasound-based method involves placement of a

Doppler probe (a combination of an ultrasound emitter and receiver along with logic to determine the frequency shift between the emitted and received sound) over the brachial artery of patients with faint Korotkoff sounds in an attempt to measure systolic BP, and also record the "ankle-arm index", which is a comparison of the systolic BP in the brachial and posterior tibial artery to ascertain the index of peripheral arterial disease (Pickering et al., 2005).

From the above descriptions, it may be noted that the BP monitoring devices containing mechanical parts or relying on inflatable cuff-based mechanisms incur an accuracy penalty. Thus there is a need for one or more alternatives to the standard BP monitoring techniques, and as mentioned earlier, PWV could be a potential answer. The following sub-section discusses all the aspects of PWV relevant to this study and the proposed approach.

2.2   Pulse Wave Velocity

As mentioned earlier, PWV differs at different regions of the arterial tree (Harada et al., 2002), so the points at which measurements are taken are an important factor to consider. The most commonly used locations, believed to provide the most reliable readings, include the aortic, common carotid, femoral, brachial, and tibial arteries, as shown in Fig. 2.1 through 2.3 (Padilla, Berjano, Saiz, Rodriguez & Facila, 2009; Harada et al., 2002; Kim et al., 2007; Salvi et al., 2008; Eriksson et al., 2002; Baguet et al., 2003; Wilkinson et al., 2010).
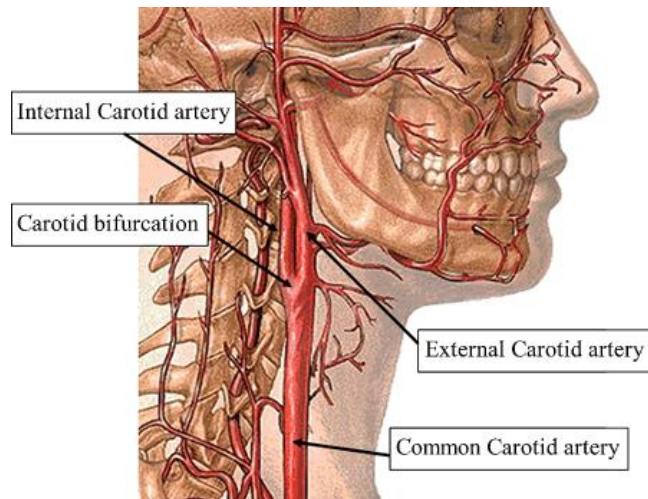
Fig. 2.1 – Carotid arteries (neck)
("Vascular Diagrams", n.d., Carotid artery anatomy)
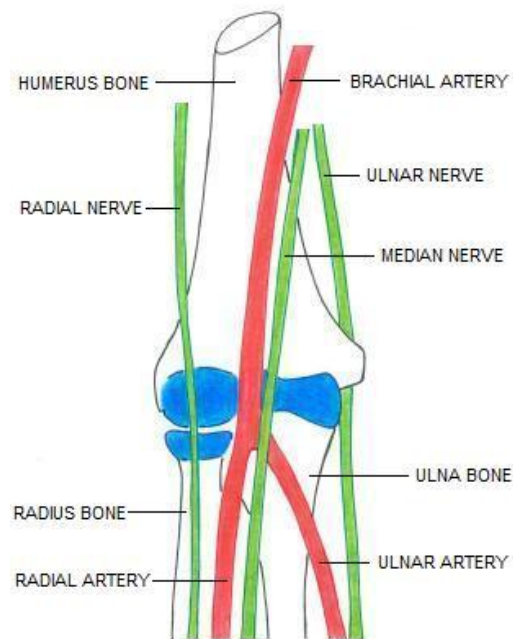
Fig. 2.2 – Brachial artery (arm)
("Elbow Anatomy", n.d., Nerves and Arteries around elbow joint)
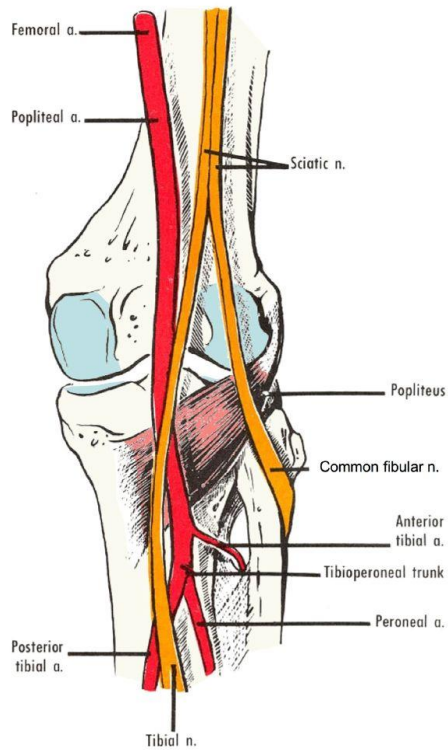
Fig. 2.3 – Femoral and Posterior Tibial arteries (leg)
("Basic Human Anatomy", n.d., Figure 15-10)

The PWV for a given region of the arterial tree can be measured by one of several different methods, of which the most common ones are described below.

1) *Arterial Tonometry* makes use of a tonometer, a device which measures pressure within the body. The device consists of non-invasive pressure sensors that are applied over the carotid and femoral arteries simultaneously (Jiang, Liu, McNeill & Chowienczyk, 2008) in order to measure the pulse transit time (PTT) "between the upstroke of carotid and femoral pulse waveforms" (Jiang et al., 2008). Once the PTT has been obtained, the distance between the sternal notch and the point of measurement over the femoral artery is measured to determine the length (L) of the arterial segment under

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observation. This can be done with the aid of a simple tape measure. With the PTT and L values in hand, the PWV may then be calculated by simply dividing the artery length by the PTT value. Commercial arterial tonometry systems which measure PWV and other arterial parameters include their own software, which perform the necessary steps to achieve the required results via inbuilt algorithms (Jiang et al., 2008).

2) *Photoplethysmography (PPG)* is a technique used to detect blood volume pulsations in the body with the aid of infrared light, as shown in Fig. 2.4 (Spigulis, Erts, Nikiforovs & Kipge, 2008). It is based on the analysis of the optical properties of a selected area of skin on the body (Kirtley, 2002). To achieve this purpose, infrared light rays are emitted into the skin at a chosen point. Depending on the blood volume in the skin, a certain amount of the light is absorbed, and a certain amount is reflected back. The change in the reflected or back-scattered light over time corresponds directly to the changes in blood volume. Thus, the point in time when the pulse wave reaches the measurement point can be determined. By employing a pair of such sensors (Fig. 2.5), the amount of time taken by the pulse wave to traverse the arterial segment under observation can be determined. This information, combined with the length of the segment, can be used to calculate PWV. PPG devices consist of an infrared light emitting diode (LED), a phototransistor capable of receiving light waves, a strip chart recorder, and a power module (Lee, Barnett, Shanfield & Anzel, 1990). The LED emits infrared light waves which penetrate the skin, and are eventually reflected back by the underlying tissue. The phototransistor then detects the reflected light, and the signal is passed to the chart recorder from which a hardcopy of the PPG

waveform is obtained. Modern, more advanced versions of these simply make use of a digital display to output the waveform.
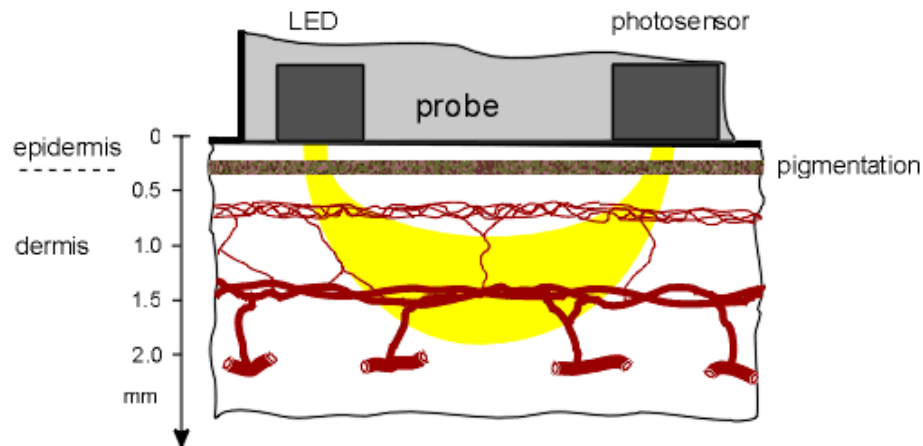
Fig. 2.4 – Infrared light emission and detection by a PPG probe
(Kirtley, 2002)

Fig. 2.5 – A pair of finger PPG sensors (Kirtley, 2002), wherein one is clipped to the top segment of the finger, and the other attached at the base of the finger. By measuring the transit time and the distance between the two points, PWV can be estimated.

3) *Doppler Techniques* make use of ultrasound principles (Fig. 2.6) by means of various types of ultrasonic transducers or sensors. As discussed in section 2.1, Doppler ultrasound can be used to detect blood movement in response to an

[21]

inflatable cuff. It can also be used to detect the pulse wave and thus provide a reference point for PWV measurement. More specifically, PWV measurement can be achieved by recording "the transit time between Doppler velocity signals in the carotid and femoral arteries" (Jiang et al., 2008). Commercial ultrasound devices provide pulsed wave Doppler velocity waveforms which can allow for PWV measurement as part of a standard vascular screening. Apart from being used to measure PWV, ultrasound techniques can also be utilised to study "intima-media thickening", which is an important indicator of subclinical arteriosclerosis and cardiovascular disease (Jiang et al., 2008). From a comparative study between arterial tonometry and Doppler ultrasound, they conclude that the flow velocity waveforms derived from pulsed wave Doppler ultrasound "can be used to measure PWV with similar reproducibility to that achieved with a standard tonometer system" (Jiang et al., 2008).
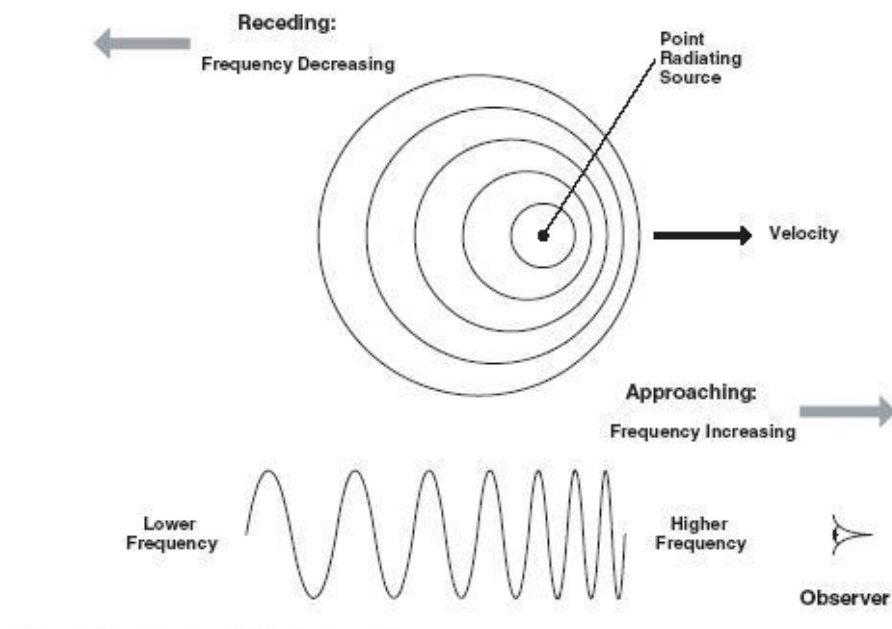
Fig. 2.6 – Doppler effect due to motion
("Aircraft Radar Basic Principle", n.d., Figure 3.7)

[22]

4) *The Electrocardiogram (ECG)* measures the electrical activity in the heart to allow the determination of any abnormalities in its functioning ("Electrocardiogram (ECG) and high blood pressure", n.d.). The electrical activity is displayed as a waveform either on a digital monitor or on a moving strip chart, and can be monitored by a medical professional to determine any unusual activity such as dysrhythmia. An ECG does not directly measure BP or PWV, but is a useful method to check if any cardiovascular or blood vessel damage has resulted from high BP. It has also been used in a number of studies alongside the aforementioned techniques as an aid in the determination of PWV, by providing a timing reference point, i.e. when blood is pumped from the heart, for the calculation of PTT (Yan & Zhang, 2005; Bolanos et al., 2004; Baguet et al., 2003; Franchi et al., 1996; Shriram et al., 2010; Jiang et al., 2008).

Each of the methods described above can be applied to one or more locations on the body to measure PWV, depending on the apparatus used. This in turn may affect the algorithm or equations used to compute PWV. The remainder of this section presents a few of the commonly used PWV equations:

- For cases wherein the PTT between two distinct points along the arterial tree and the length (L) of the arterial segment between those points have been obtained, PWV can be calculated by simply dividing the two values, as illustrated by Jiang et al. (2008):

  PWV = L / PTT   [Eqn. 2.1]

The transit time is usually measured between two points along the arterial tree, and any of the above methods or combinations of methods may be employed at each site to determine the arrival of the pulse wave. An example of a single technique could be the use of a tonometer at each of the points, and a combination of techniques could be an ECG and a PPG reading.

- One of the most common, established relationships to have been employed in a number of studies (Hermeling, Reesink, Reneman & Hoeks, 2007; Lim & Lip, 2008; Shao, Fei & Kraft, 2004) to calculate PWV is the Moens-Korteweg formula, which relates PWV directly to the stiffness of the arterial wall, and is given as:

$$PWV \quad = \quad \sqrt{(Eh/2)pr} \qquad\qquad \text{[Eqn. 2.2]}$$

where E is the elastic modulus of the arterial wall (known as Young's Modulus), h is the thickness of the arterial wall, p is the density of blood, and r is the vessel radius at the end diastole.

- Franchi et al. (1996) evaluated a much more complex mathematical model, based on Womersley's expression for PWV for a single arterial segment:

$$PWV \quad = \quad \sqrt{\frac{M_{10}\pi r_2}{\rho|C|} \cdot \frac{1}{\cos\Psi}} \qquad\qquad \text{[Eqn. 2.3]}$$

$$M_{10} \quad = \quad \left|1 \; - \; \frac{2J_1(i^{3/2}\alpha)}{i^{3/2}\alpha J_0(i^{3/2}\alpha)}\right| \qquad \text{and} \qquad \alpha \; = \; r\sqrt{\frac{\omega\rho}{\mu}}$$

where ρ refers to blood density, C is the arterial compliance, r is the arterial radius, μ is blood viscosity, Ψ is the dissipation angle, ω is the angular frequency of oscillations, and J_0 and J_1 are the Bessel's functions of first type and order 0 and 1 (Franchi et al., 1996).

- Harada et al. (2002) take a different approach and propose a means by which PWV may be ascertained from a "one-point" measurement, as opposed to obtaining signals from two points of an arterial segment. The method makes use of the Water-Hammer equation for forward travelling waves:

$$PWV \quad = \quad (\Delta P / \Delta V)/\rho \qquad\qquad\qquad\qquad [Eqn. 2.4]$$

where ΔP refers to the change in BP and ΔV to the change in velocity over a period of time, "at a fixed point in an artery, caused by a forward wave travelling from the heart to the periphery" (Harada et al., 2002). ρ is the blood density. It is assumed that there are no effects of backward travelling waves, i.e. waves travelling from the periphery to the heart, during measurement.

PWV has proven to be an important physiological measure in the monitoring of cardiovascular and arterial health. The fact that it directly relates to BP has been closely investigated in a patent filing by Sullivan, Cheung, Sullivan, and Wise (2006), wherein they use PWV to derive systolic and diastolic BP values. Though out of scope for this study, their methods deserve mention, as literature on PWV to BP conversion is relatively scarce, and it may be possible in the future to adopt their technique and measure BP directly from PWV using the approach presented in this study.

2.2.1 Calculating Blood Pressure from Pulse Wave Velocity

Sullivan et al. (2006) developed a system for passive physiological monitoring, consisting of a variety of sensors and apparatus to acquire different signals, and a computer to process them into meaningful information. The system analysed a variety of mechanical, thermal and acoustic signals, which reflected cardiac functioning, respiration, pulse, apnea, and temperature, with data being recorded and displayed in realtime.

In this patent publication, the inventors describe how the systolic and diastolic BP values may be computed from PWV with the aid of the Bernoulli Equation; a well-known equation in the field of fluid dynamics, derived from Newtonian mechanics and the principle of conservation of energy. They introduce the following version of the equation that includes systolic BP and diastolic BP:

$$P_{AVG} = P_D + 1/3 * (P_S + P_D) \hspace{3cm} \text{[Eqn. 2.5]}$$

Where, P_S refers to systolic pressure, P_D refers to diastolic pressure, and P_{AVG} refers to the average pressure, given by the following equation:

$$P_{AVG} = \rho g h + 1/2 * \rho * V^2 \hspace{3cm} \text{[Eqn. 2.6]}$$

where ρ is the density of the fluid (blood, in this case), g represents the gravitational constant, V refers to velocity, and h refers to the height or head energy term.

The human cardiovascular system may be modelled as a network of pipes and valves, similar to that of a household plumbing system. This model allows the use of the Water Hammer equation mentioned earlier on in this section, to represent the speed of the pulse wave:

$$c = 1/\rho * \Delta P/\Delta V \qquad\qquad\qquad [Eqn. 2.7]$$

where c is the speed of the wave, ρ is the fluid density, and ΔP and ΔV refer to change in pressure and change in velocity, respectively.

Once velocity (V) has been determined, the following equations derived from all of the above can be used to give the final values of systolic BP (P_S) and diastolic BP (P_D):

$$P_D = \tfrac{1}{2} * P_{AVG} * V^2 - \rho * c * \Delta V \qquad\qquad [Eqn. 2.8]$$
$$P_S = P_D + \rho * c * \Delta V \qquad\qquad\qquad [Eqn. 2.9]$$

Thus by utilising a combination of the Water Hammer and Bernoulli equations, a BP reading can be arrived at based on PWV using [Eqn. 2.8] and [Eqn. 2.9].

The final area investigated in the PWV domain relates to examples of some different types of devices that were developed as part of various studies to measure PWV. These are discussed in the following sub-section.

2.2.2 Existing physiological systems using Pulse Wave Velocity

Williams, Jones & Doughty (1997) developed a wearable sensor sleeve for the measurement of BP to detect preeclampsia (pregnancy-induced hypertension). Their device made use of the PPG technique, via a wide-angle infrared LED and low power photodetectors, to calculate PWV for the determination of BP. The sleeve was worn around the patient's forearm, with one end lying just below the elbow, and the other a little above the wrist, as the authors discovered that "by fixing one detection point 2 cm from the fold of the elbow, a good pulse could be found in all subjects...irrespective of their physical characteristics" (Williams et al., 1997).

McCombie, Shaltis, Reisner and Asada (2007) proposed and developed a new method of calibration for a wearable autonomous BP monitoring device, known as the adaptive hydrostatic BP calibration algorithm. Their method made use of a wrist watch sensor device, which included two photoplethysmograph sensors (one placed at the wrist and the other against the little finger of the left hand), a fluid-filled tube, and a pressure sensor. The study also showed how this novel calibration technique could be applied to PWV measurements obtained from the BP monitoring device.

Eriksson et al. (2002) described a new, non-invasive ultrasonic technique to measure PWV using "a new colour Doppler modality for measuring tissue motion" in their investigation of how system sensitivity and resolution may affect PWV estimation. The study made use of the HDI-5000 ultrasonic system provided by ATL Ultrasound, a US manufacturer of ultrasound systems, to gather the tissue Doppler imaging (TDI) information required to calculate PWV.

PWV has previously been established as a strong indicator of vessel stiffness (Shao, Fei & Kraft, 2004), but it also "has good reproducibility, requires little technical expertise, and there is evidence from a number of large prospective independent studies to support its use as an independent predictor of vascular outcomes" (Korpas et al., 2009). Analysis and measurement of PWV is thus considered beneficial and important in various clinical studies, and has also garnered significant research interest due to having obvious benefits.

2.3   Telemedicine Systems and Devices

Since this study already makes use of a mobile phone to acquire data for PWV calculation, one of the future possibilities is that the methods and algorithms presented here may be ported over to function completely on the mobile phone itself, effectively producing a telemedicine system for PWV estimation. Hence, the field of telemedicine is also examined briefly as part of this investigation.

Telemedicine refers to the analysis of patient information by a medical professional who is geographically separated from the patient, for the monitoring and treatment of one or more chronic ailments (Murdoch, 1999). It involves the transmission of medical data from one point to another via existing telecommunication infrastructure. It offers the advantages of constant in-home monitoring and cost effectiveness to patients with chronic diseases. The concept of telemedicine is not new, as an example cited by Murdoch shows; William Einthoven, the inventor of the ECG, made attempts to transmit ECG data over telephone lines as early as 1906.

One of the challenges of telemedicine is developing the device responsible for performing patient monitoring; it must ideally be non-intrusive, comfortable, and must not restrict patient mobility. Scheffler and Hirt (2005) state that "the development of wearable medical devices has to take additional user requirements into account, when compared with the design of stationary equipment". They proposed the following guidelines for the development of wearable telemedicine devices:

- Physical dimensions need to be kept to a minimum; 60 x 50 x 15 cm^3 to suit the size of a forearm, for example. This requires many mechanical and electrical considerations, both before and during the development stages. The device must not be bulky and heavyweight, as this becomes rather cumbersome if the patient is to undergo ambulatory monitoring.

- Operating time must be for 24 hours at the very least without having to recharge the battery. The device should use low power components wherever possible, or a secondary rechargeable battery if there are high power requirements.

- In order to comply with various health insurance reimbursement schemes, the device must exhibit reliability and a minimum four year field life. It must be sturdy and impervious to small shocks.

- Methods for data transmission should be carefully chosen after weighing their pros and cons; a wired setup will pose mechanical issues and limit portability, whereas a wireless solution will consume significantly more power.

- The type of sensor to be used to acquire signals from the user's body needs careful consideration, as it may not fit in the desired casing or may be difficult to position in a way that ensures consistent contact with the user's skin.

A number of studies have been carried out in this field, each one targeting various health conditions and proposing different solutions. A short review of some such studies follows.

Williams et al. (1997) proposed and developed a telemedicine system for the detection of preeclampsia. Their study included the construction of a sensor sleeve worn on the forearm, which made use of PPG sensors to record PTT of the signals. The system also incorporated the MeterLink ringless telemetry system, courtesy of British Telecom, to relay the measured data to the required hospital's computers by polling the sensors at frequent intervals without any patient intervention. The system functions as follows:

- The hospital computer dials the patient using the ringless protocol.

- The call is picked up by a telephone interface unit (TIU), which then sends a signal to the sensor sleeve, causing it to activate the sensors.

- Data in the form of 10 kHz bursts is transmitted to the TIU for a period of 20 seconds.

- The TIU relays both the number of bursts and the total number of 10 kHz pulses to the hospital computer.

[31]

- The computer then calculates heart rate, PTT, and BP wave velocity.

- The data and time of recording are stored and compared to previous readings.

Scheffler and Hirt (2005) examined two wearable devices for telemedicine applications:

- The Advanced care and alert portable telemedical MONitor (AMON) is a wrist-wearable monitor for heart patients. The device contains multiple sensors to measure a variety of physiological parameters of the patient, including temperature, BP, blood oxygen level, and ECG. Sensor data are relayed to a medical care centre via the Global System for Mobile communications (GSM).

- The Cubic Belt-Integrated Computer (QBIC) is a generic computing platform embedded in a standard belt for home and hospital use. It features low power consumption, wireless capabilities, and data storage. An example of its usage is mobile ECG recording, whereby the device can collect and process signals from ECG sensors, and then save the data to a removable storage device or transmit it over a wireless network.

Schott (2002) describes an external automatic wearable defibrillator developed by Lifecor in the USA. The system, designed to prevent death in the event of a cardiac arrest or life-threatening dysrhythmia, contains a number of components placed within a wearable cotton vest that is secured around the patient's chest area. There are four electrode sensors located within an electrode belt, one each on the front and sides of the chest, and one on the back; these are responsible for continuously monitoring heart rhythm. The electrode belt also contains three

therapy pads which deliver the life-saving electric shocks to the patient. Two of these lie vertically against the patient's back, and the third lies horizontally on the left side of the chest. The electrode belt is connected to a monitor which contains a signal processing computer, batteries, and capacitors. In the event of irregular heart rhythm being detected, the monitor emits an audible tone to alert the patient that a shock will be delivered within a minute. The patient can dismiss the alarm and avoid receiving the shock if it is wrongly detected via buttons placed on the monitor, but if no response is made, the electric shock is delivered. The device records the patient's ECG during the process, up to 15 seconds after normal heart rhythm has been restored. This data is stored and can then be transmitted via the included modem to a secure website for use by medical professionals. Clinical studies on wearable defibrillators reported a drop of 30% – 40% in mortality rates due to dysrhythmia.

Bolanos et al. (2004) developed a non-invasive and portable Personal Digital Assistant (PDA) based BP and ECG monitoring system capable of recording ECG and BP waveforms, and calculating heart rate, systolic BP, and diastolic BP. The device could then transmit the data wirelessly to the PDA for storage on flash memory or direct display on the screen. ECG signals were acquired using a custom-built low power ECG amplifier module, and BP waveforms were procured by performing finger PPG. The device was capable of continuous operation for a 24 hour period.

In their research on chronic disease management, Trudel et al. (2007) discuss the "development and testing of a novel approach to tele-monitoring for improved chronic disease management". They developed a mobile phone based system for the monitoring and control of high BP in type II diabetics, by merging together

several existing technologies and devices to form a complete system. Patient BP was recorded using a standard Bluetooth enabled cuff-based home monitor. The data is relayed via Bluetooth to the mobile phone running a custom software application, which then stores and also transmits the data securely using a Secure Socket Layer (SSL) connection to a central data repository, via the General Packet Radio Service (GPRS) network. Clinical rules were automatically applied to the incoming data at the central server, and alerts generated and conveyed to both the patient and his or her doctor, in the event of abnormal readings. The system was also capable of issuing alerts and reminders to patients to ensure that they adhere to their measurement schedule. It thus enabled timelier measurements and appropriate follow up of the patient's care.

Telemedicine devices typically offer patients of chronic ailments the advantage of continuous ambulatory monitoring, or at the very least, the means to check on various vital signs while on the move, at any time and place. This promotes a sense of security to a certain degree, and can also help in early detection of any abnormalities, which can be used to subsequently diagnose problems at the earliest possible time.

2.4   Summary

This section presented the knowledge uncovered from investigating various sources of current and past literature in the field of PWV, BP and telemedicine. Though PWV is the central focus of this study, it shares an important relation with BP, and the proposed approach in this study holds potential as a future telemedicine system, hence the inclusion of these two topics as well. To

summarise, the main points of interest derived from examining the literature are listed below:

1) Traditional methods of BP measurement include the Korotkoff method, employed by manual sphygmomanometers, and the oscillometric technique, present in the digital automated monitors.

2) PWV is known to have a positive correlation with BP, however literature detailing the possibility and methods of deriving BP from PWV is scarcely available. The one source identified to contain this information combines the Bernoulli Equation and Water Hammer Equation to derive systolic and diastolic BP from a PWV reading.

3) There exist several techniques for PWV measurement, but the devices implementing these are mostly unattainable to the common individual due to their complexity and price. Thus PWV measurement is limited only to clinics or hospitals, with no real home monitoring options.

4) Different studies present different equations for the computation of PWV, which depend on the type of devices being used and the arterial segment under observation.

5) PWV has garnered significant research interest over the years, evidenced by numerous studies in the field and attempts at creating new, more accessible devices for its measurement.

6) The field of telemedicine and mobile healthcare is continuously advancing, especially with the increasing power of new generation mobile phones. Telemedicine devices can help reduce mortality by enabling timely detection of physiological anomalies within the body, leading to early diagnosis and treatment.

3.0 RESEARCH METHODS AND MATERIALS

This chapter provides information about the specific research method adopted by the study, detailing the steps involved in it as well as describing the processes that relate the method to this study. The list of materials and instruments required to carry out the proposed work is also detailed, and lastly, an overview of the proposed approach is presented in brief detail.

3.1 Research Methodology

This study adopts an engineering approach, as summarised in Fig 3.1. The approach consists of the following steps:

1) *Identify and examine existing solutions related to the study:* This relates to the literature review stage, where various past and present studies related to PWV measurement were examined, the findings of which were collated and presented in Chapter 2.

2) *Propose a new, unique solution:* Based on the review of existing methods and findings, a new approach for PWV measurement based on mobile phone audio and video sensors and Doppler ultrasound was formualted.

3) *Develop and refine the proposed solution:* This involved the various processes required to design and implement a functional prototype that could demonstrate the usability and working of the proposed approach. Different

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experiments were carried out to test various factors, in order to develop an appropriate algorithm.

4) *Test and evaluate the proposed solution:* Once the prototype was built, it was subject to testing via some more experiments so that its results could be evaluated against existing standards to gauge its efficiency and correctness.

5) *Reiterate through the developmental and evaluation phases to improve the proposed solution until the best possible one is achieved:* This refers to the recursive process of going back to certain points in the developmental phase and making changes to any aspects that were found to negatively impact upon the final results, so as to achieve the best possible outcome.
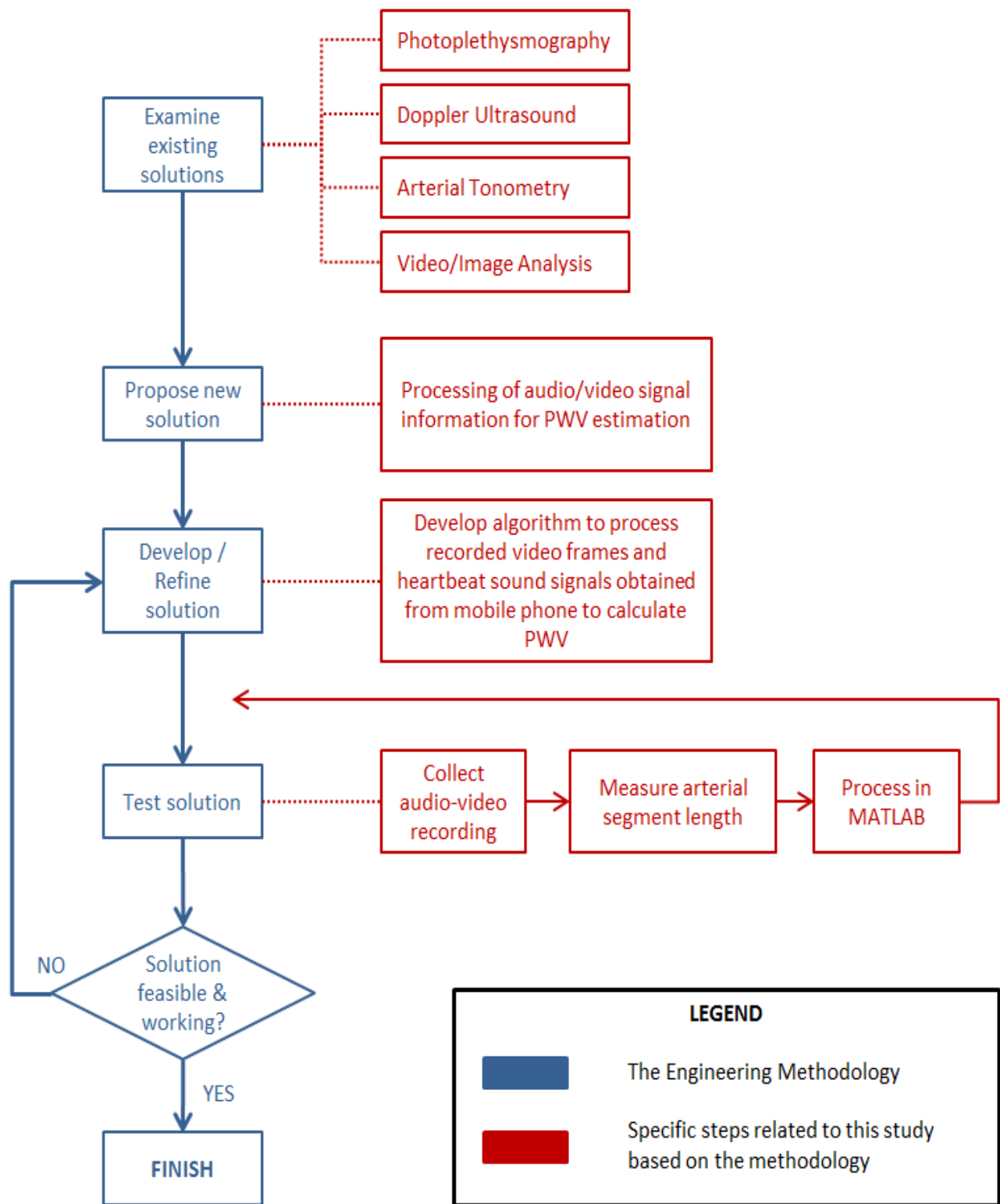
Fig. 3.1 – The engineering research methodology and its application to the study

3.2   Materials and Instruments

To undertake the implementation of the proposed approach, the following materials were used:

1) Mobile smartphone with an embedded camera capable of video recording, and microphone input (iPhone 3GS, iPhone 4, and HTC Desire).

2) Doppler ultrasound probe (2 MHz "Baby Sound A" fetal Doppler unit manufactured by Contec Medical Systems Co. Ltd., China).

3) Stereo cable (3.5 mm to 3.5 mm plug).

4) 1/8" Microphone input adapter (3.5 mm 4-conductor TRRS male to 3.5 mm mic input jack).

5) Stereo headphones.

6) Computer running the Microsoft Windows operating system.

7) Software (MATLAB and Xilisoft Video Converter.

8) Commercial Heart Rate monitor (Polar T31).

9) Transmission Gel.

10) Tape Measure.

Fig. 3.2 – Instruments required for audio and video data acquisition

3.3   Approach

We made use of two of the mobile phone's sensors, namely the camera and microphone input, to provide reference points for PWV measurement. The microphone was used to listen for the heartbeat via Doppler ultrasound and provide the first reference point. The camera was used to mimic a photoplethysmograph sensor, similar to the implementation of Poh, McDuff, & Picard (2010), giving the second point of reference. A detailed breakdown of the approach is as follows:

1) The mobile phone is used to capture a video recording of a region of interest on the user's body (eg: forehead or palm). During this process, the Doppler

[41]

probe is plugged into the phone's microphone input, and placed on the left side of the user's chest to record heartbeat. This results in the output of a multimedia file. Also, the distance between the two sites of measurement is measured via a tape measure.

2) For ease of prototyping, analysis and processing was carried out in MATLAB. Some pre-processing was carried out on the audio and video data to convert them into useful signals, and these were then analysed manually to first verify their usability. Following this, a custom MATLAB program was developed to automatically calculate PWV, given the two signals and the length of the arterial segment. This involved a number of smaller steps, which have been fully detailed in Chapter 5.

3) The developed algorithm was then tested using additional video recordings from members of the research team under some different conditions (at rest and after some physical activity).

4) Results from the experiments were documented and compared against reference values and other results found in relevant literature to determine their validity and usefulness, as a commercial PWV monitor was not available during this study.

5) Once the results were found to be satisfactory and consistent, the study was brought to a conclusion.

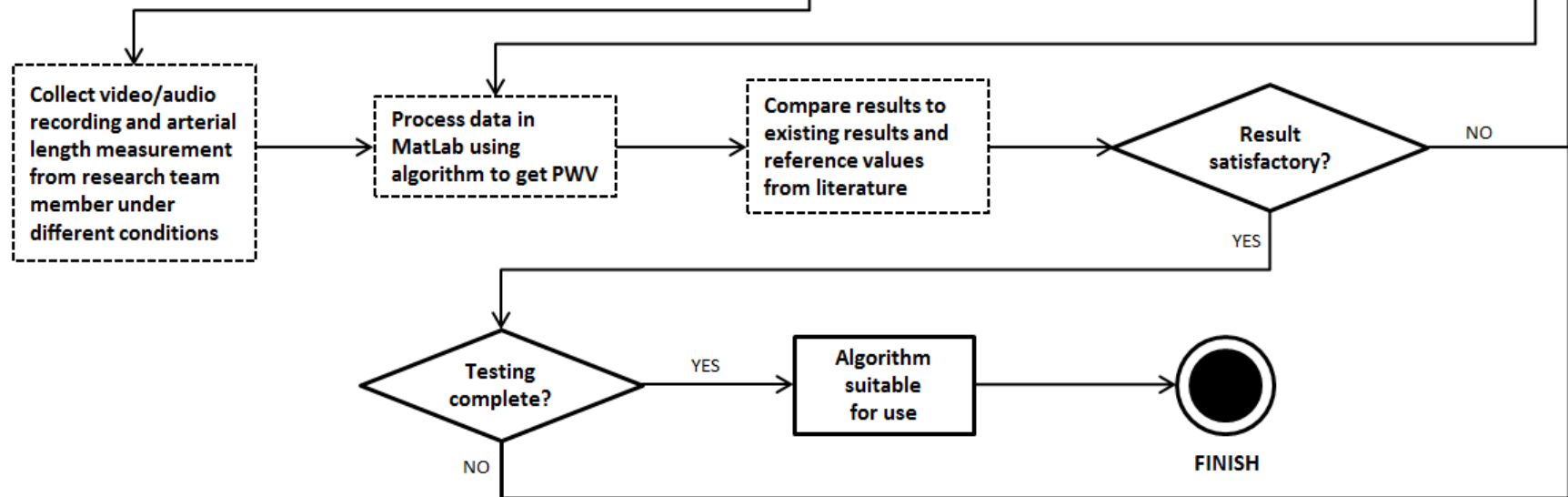The process described here is briefly summarised by Fig. 3.3:
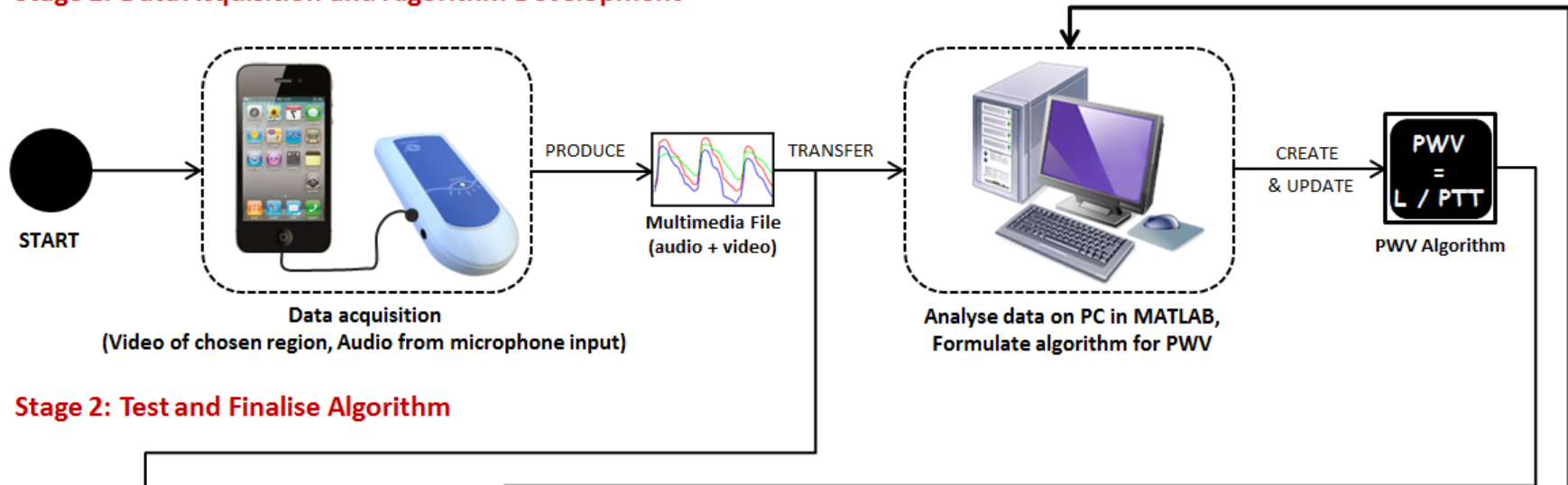
Fig. 3.3 – Summary of the proposed approach

[43]

4.0   DATA ACQUISITION AND EXPERIMENTAL SETUP

This chapter provides details on all aspects pertaining to the data collection process. The proposed approach deals with audio and video data acquired from the appropriate mobile phone sensors, i.e. audio input and camera. More specifically, the following are required:

1) An audio recording of the subject's heartbeat from the chest. This was acquired with the aid of the portable Doppler probe connected to the mobile phone's audio input.

2) A video recording of a pre-determined area on the subject's body, hereafter referred to as the region of interest, acquired using the mobile phone's camera.

Investigation into relevant literature reveals multiple methods for PWV measurement, some of which were outlined in Section 2.2. The proposed approach involving the use of audio and video data is similar to the two-point estimation techniques using PPG and ECG signals employed in other studies (Liu, Hsu, Chen & Wu, 2010; Yoon, Cho & Yoon, 2008). The audio recording gives the first reference point, i.e. the time when blood is pumped from the heart, and the video recording of the region of interest signals the arrival of the blood in the vessels or capillaries of that area. This data is then processed and analysed on a computer in MATLAB to calculate the PWV. Details on this process and the algorithms used are discussed in Chapter 5.

The sub-sections that follow present the details of video and audio data acquisition, as well as all the experiments carried out as part of the investigation. Experiments were performed in order to:

1) Determine which of the three mobile phones possessed the hardware required for the best possible results.

2) Narrow down the choice of region of interest for video acquisition from four possible regions.

3) Confirm that the results obtained from video acquisition using the chosen mobile phone and region of interest from the prior experiments were valid.

4) Confirm that the results obtained from audio acquisition using the chosen mobile phone and Doppler ultrasound probe were valid.

To achieve these goals, a commercial heart rate monitor (Polar T31, Polar USA) was used to log the subject's heart rate in each of the experiments, and this was then compared to the heart rates calculated manually from the processed audio and video signals. Visual inspection of the signal waveforms was also useful in determining its usability to a certain extent.

4.1   Video Data Acquisition

The video data is to be acquired by capturing a recording of a specific location on the body using the mobile phone camera. For this, there are multiple possible locations that could potentially be used, so part of the investigation involved narrowing down the selection to the one region that could be easily accessed by the user as well as provide reasonably good results. The regions chosen for examination included the forehead, the left palm, the left index finger tip, and the left thumb tip. The main difference between

[45]

these locations is that the palm and forehead regions are kept at a certain distance from the camera lens, while the tip of the thumb and index finger are to be placed directly over the lens, making firm contact with it. Consequently, this affects the quantity of light that is absorbed and reflected, which ultimately impacts the quality of the signal. Apart from the difference in their proximity to the camera lens, these regions were chosen for specific reasons. The palm contains a good supply of blood vessels, which makes colour changes due to blood flow in the skin's surface more easily noticeable than other areas. The forehead region was chosen based on the work conducted by Poh et al. (2010), in which they chose a segment of the subject's face as their region of interest. Finally, the tips of the thumb and index finger were chosen due to the fact that they are viable measurement sites for commercial pulse oximeters, and more recently, for mobile phone heart rate monitoring applications ("Instant Heart Rate", 2011).

For processing and analysis purposes, it is important that the images obtained from recording the abovementioned regions are clear and of good quality (Fig. 4.1 – a). In addition to the region itself, there are a number of other factors that could affect the results. These include lighting conditions, video resolution and clarity, distance between the camera and region of interest, and motion artifacts. If one or more of these are not setup properly, the acquired video data may be rendered useless. For example – inadequate lighting will provide a poor image overall; bringing the camera too close to the region of interest (in the case of the non-contact regions) will result in loss of focus and a blurry image; and using too low a resolution could result in pixelated video output – all of which can negatively affect the data collection. If the recorded video is plagued by one or more of these issues, the signal resulting from the processing will not be suitable for analysis as it may contain undesirable noise or other artifacts (Fig 4.1 – b).
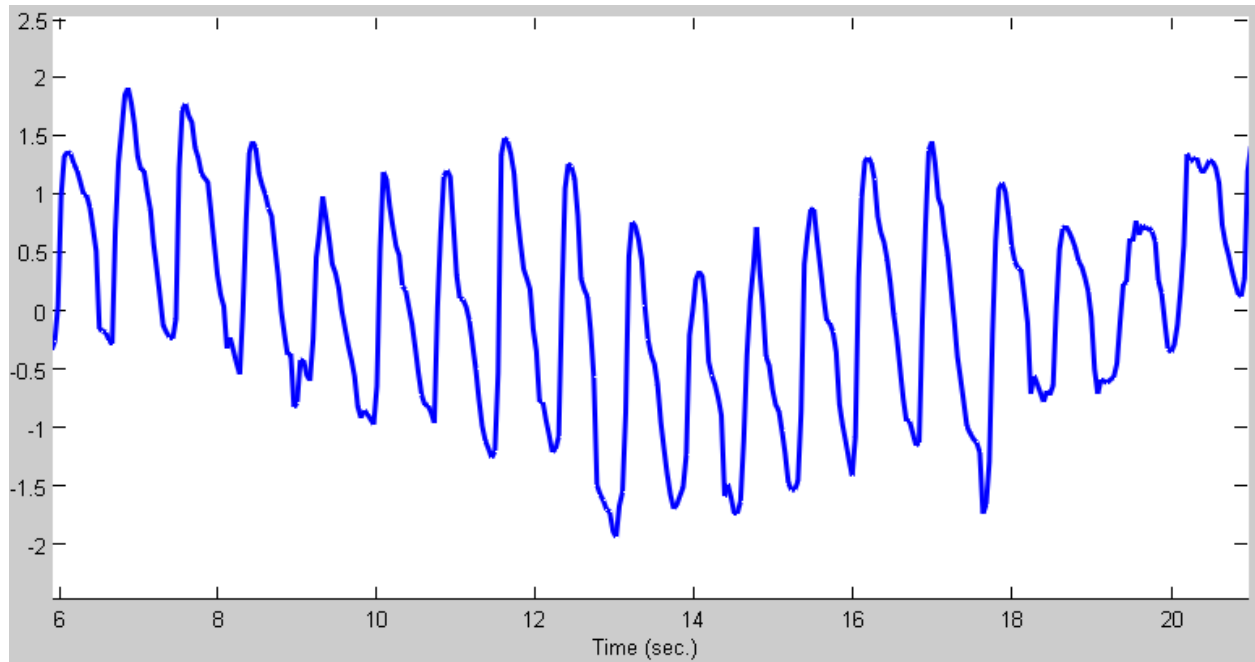
(a) A clean video signal
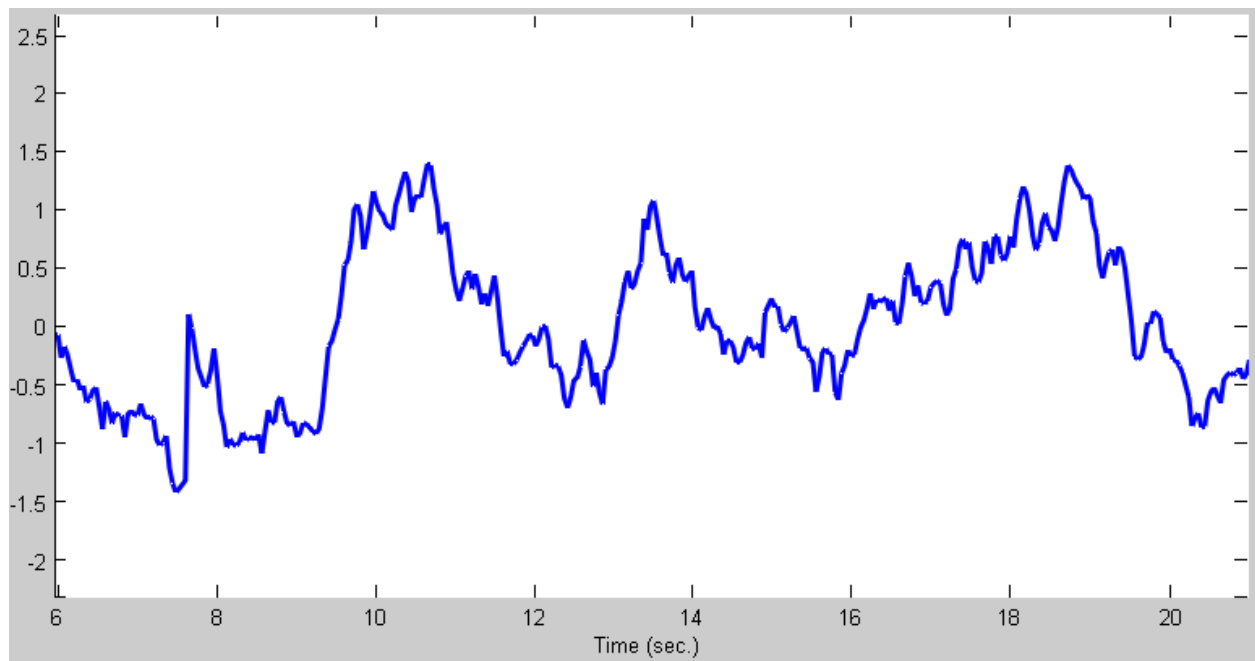
(b) A noisy and distorted video signal

Fig. 4.1 – Examples of waveforms produced after processing video data

Based on the work carried out by Poh et al. (2010), a custom MATLAB program was developed to process the video signals and convert them into graphical waveforms. For this, a sliding window was first used to extract a 30 second block of data from the whole signal. The mean red ($x_1(t)$), green ($x_2(t)$), and blue ($x_3(t)$) intensities for each frame of the video were then extracted and stored in individual arrays. Each of these colour traces was normalised by subtracting the mean and dividing the result by the standard deviation of the set of values, as shown in [Eqn. 4.1] (Poh et al., 2010):

$$x_i'(t) = x_i(t) - \mu_i / \sigma_i \qquad\qquad \text{[Eqn. 4.1]}$$

Here, $x_i(t)$ refers to each of the individual colour traces (i = 1, 2, 3), μ_i refers to the mean of $x_i(t)$, and σ_i refers to the standard deviation of $x_i(t)$.

The resultant values could then be plotted on a graph for manual inspection, an example of which is shown in Fig. 4.2. The sliding window is then moved over to capture the next one second of new data, and this continues until the entire signal has been processed.
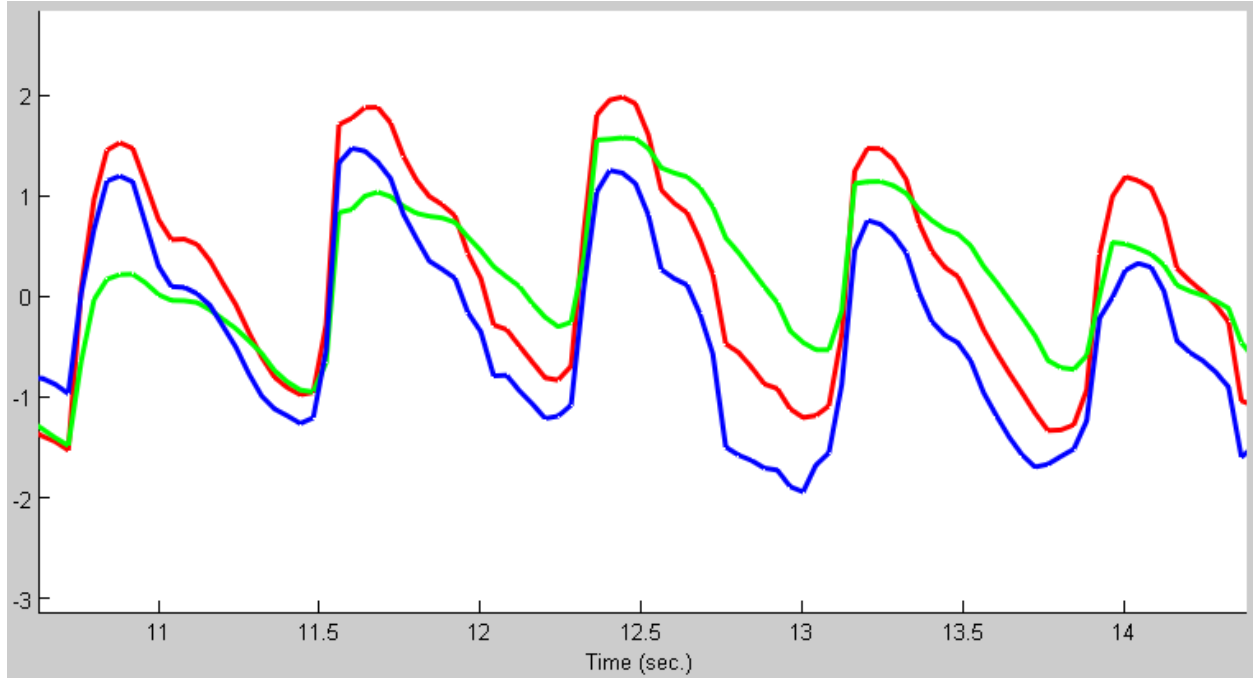
Fig. 4.2 – Section of a processed video signal showing the individual RGB channel data

As can be seen from the figure, the resulting waveform consists of distinct sets of peaks for each channel. By measuring the distance between two consecutive peaks on any one channel and applying [Eqn. 4.2] given below, heart rate can be calculated in beats per minute (bpm):

HR = (1 / (X_2 − X_1)) * 60   [Eqn. 4.2]

Where, X_1 and X_2 are the times at which two consecutive peaks occur, measured in seconds.

Fig. 4.3 illustrates manually picking the data points in MATLAB, where each point picked gives the corresponding time at which that peak occurs. This allows for measurement of the distance between consecutive peaks from the processed video signal. The blue colour channel was chosen for all the experiments, as visual inspection of the

[49]

waveforms revealed it to be the one that gave the most clear and distinct peaks as compared to the red and green channels.
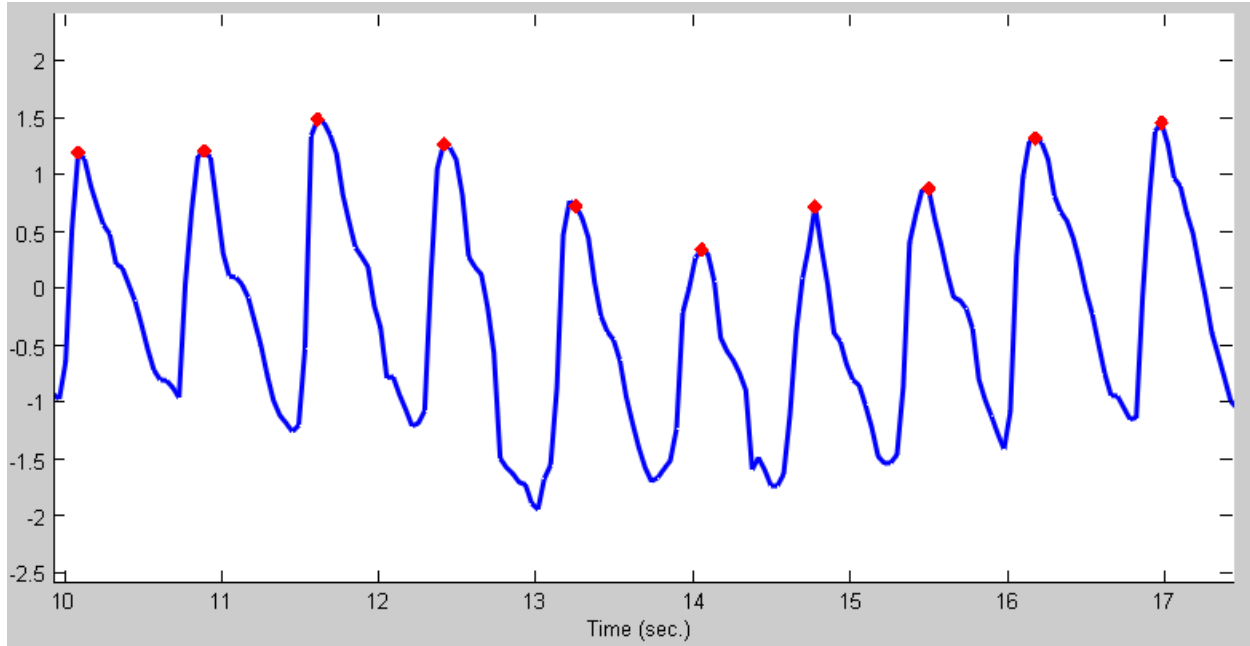
Fig. 4.3 – Data points to measure the distance between video peaks

The experiments carried out with regards to video data acquisition are detailed in the following sections. All experiments are carried out in an indoor environment.

4.1.1   Experiment 1: Selection of Mobile Phone device

<u>Aim of the Experiment</u>

For the purpose of this study, three mobile phones were acquired for testing, each with different camera hardware and capabilities. The aim is to experimentally determine which of these phones possesses the required hardware capabilities that will enable the best possible audio and video signal acquisition.

[50]

Instruments and Materials

1) Mobile Phones with camera

- iPhone 3GS (3.0 MP, 640 x 480, auto frame rate, no flash)

- HTC Desire (5.0 MP, 1280 x 720, auto frame rate, with flash)

- iPhone 4 (5.0 MP, 1280 x 720, auto frame rate, with LED flash)

2) Windows PC

3) Software

- MATLAB

- Xilisoft Video Converter 6

4) Polar T31 heart rate monitor

5) Gel

Prerequisites

Gel is applied to the Polar T31 monitoring strap and this is fitted around the subject's chest to allow for simultaneous heart rate monitoring.

Setup

For the purpose of this experiment, the tip of the left index finger was the chosen region of interest. The following steps are then performed for each of the mobile phones:

1) The subject is seated with their left arm extended forward and elevated to heart level. The arm must be supported by a suitable surface so as to reduce strain that may result in unwanted movements.

2) The mobile phone's camera application is activated and the flash is switched on. For the iPhone 3GS, sufficient ambient light was provided to compensate for the lack of a flash sensor.

3) The subject holds and positions the phone in a way that the tip of the left index finger covers the camera lens and flash adequately.

4) The subject begins the video recording on the mobile for a duration of 1 minute. During this time, the subject must breathe normally. At the same time, heart rate monitoring using the Polar T31 is carried out, with results being logged on the PC.

5) Recording is stopped at the 1 minute mark, and the video file is transferred to the PC. Monitoring with the Polar T31 is also stopped.

6) The video file is converted from the MOV format to the AVI format with a frame rate of 25 fps using Xilisoft Video Converter 6.

7) In MATLAB, the AVI file is converted to an uncompressed AVI, and then processed as described in Section 4.1.

Results

A section of the processed video signals from each of the three mobile phone recordings of the left index finger tip can be seen in Fig. 4.4. It is immediately evident from the waveforms that the signal from the iPhone 3GS (Fig. 4.4 – a) is highly noisy, having no distinct peaks or consistent structure. This completely contrasts the clean signals obtained from the HTC Desire (Fig. 4.4 – b) and iPhone 4 (Fig. 4.4 – c), in which the peaks are uniquely identifiable, and occur at regular intervals over some time period, clearly signaling the arrival of each pulse wave at the site of measurement.
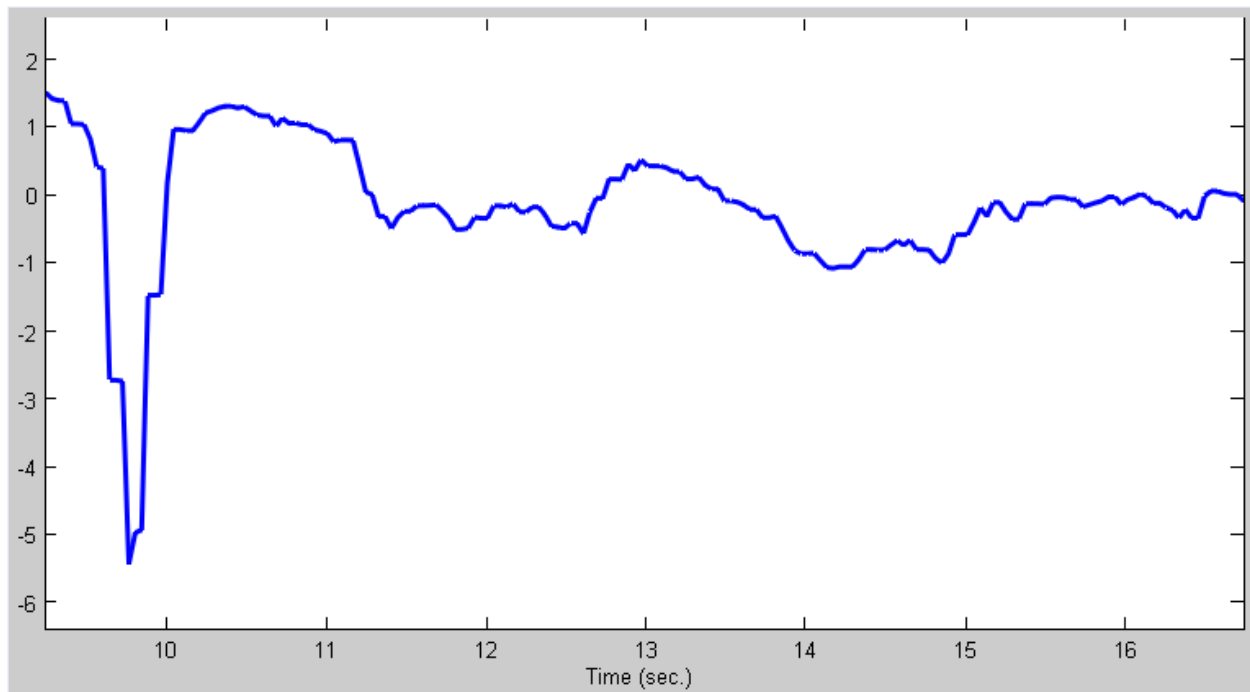
(a) iPhone 3GS signal
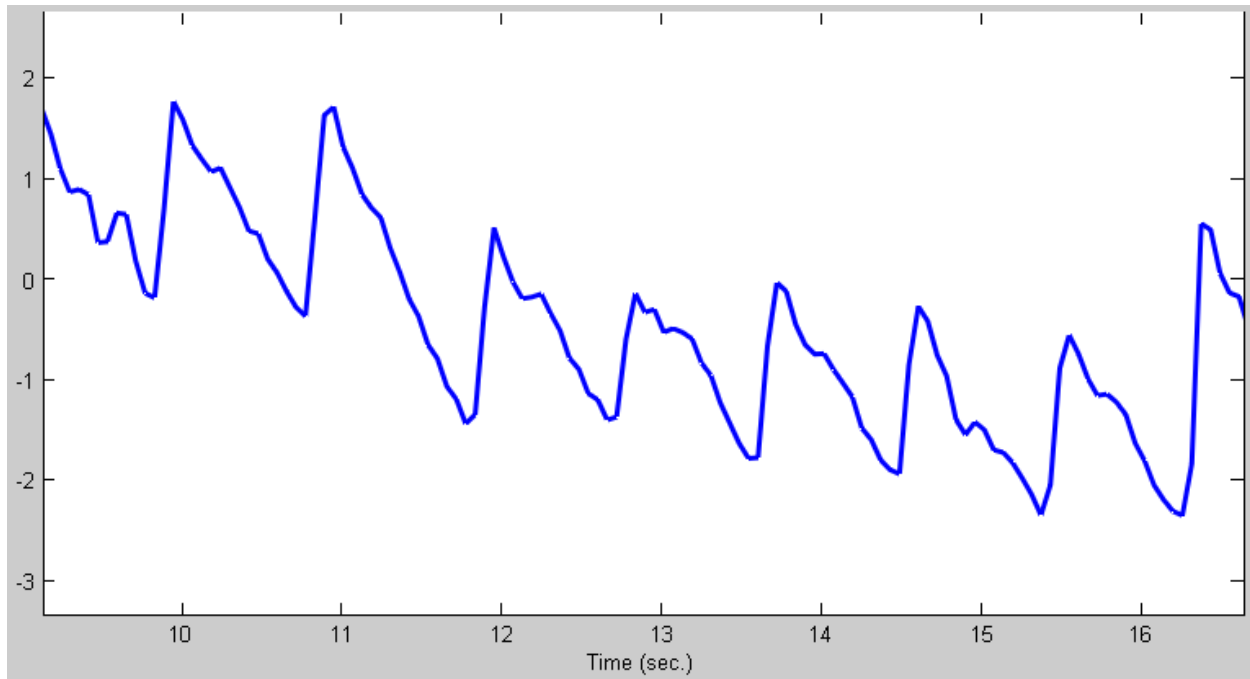
(b) HTC Desire signal
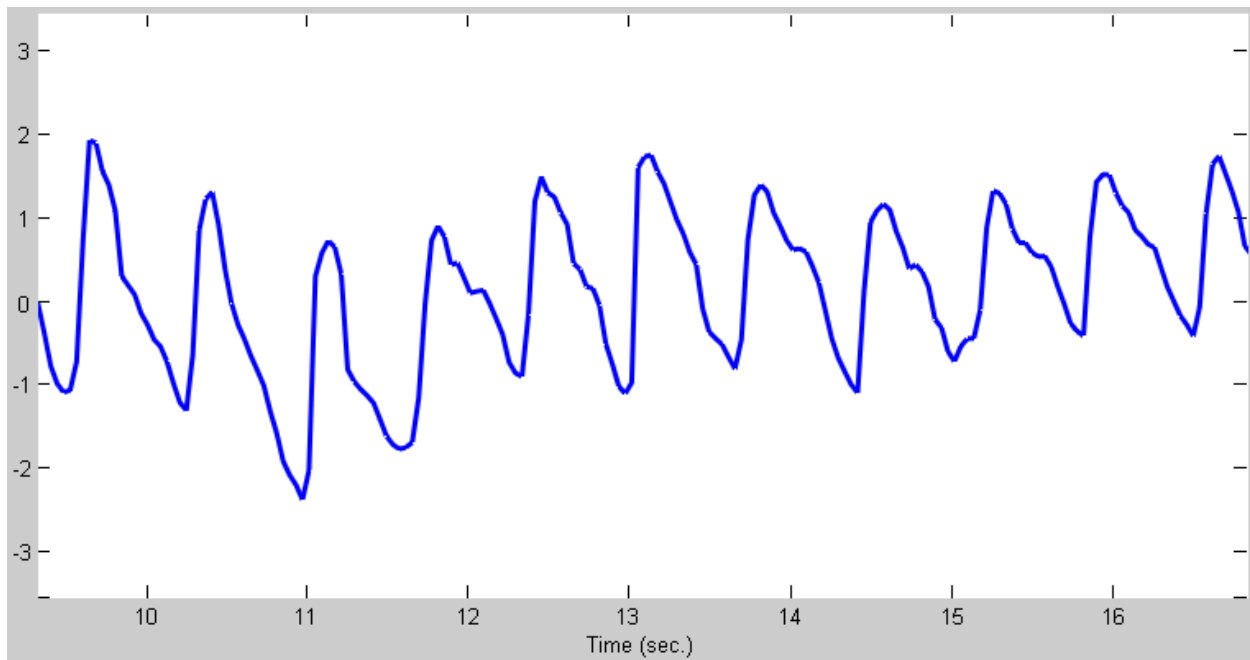
(c) iPhone 4 signal

Fig. 4.4 – Section of video waveform from a recording of the left index finger from all 3 phones

Hence, [Eqn. 4.2] can only be applied to the two clean signals for heart rate calculation, as distinct peaks are required. For this, a pair of data points from two consecutive peaks of the waveform was chosen at every 10 second interval, thereby giving a total of 6 measurements over the 60 second time period. Tables 4.1 and 4.2 compare the manually obtained results against those of the Polar T31 monitor:

Table 4.1
Heart Rate from video data of index finger tip vs. Polar T31 monitor (HTC Desire)

READING	HEART RATE (bpm) from Video Data	HEART RATE (bpm) from Polar T31	DIFFERENCE (bpm)
1	60.06	67	6.94
2	60	66	6
3	77.92	73	4.92
4	63.62	66	2.38
5	68.18	69	0.82
6	63.82	68	4.18
MEAN	65.6	68.16	4.2
STD. DEV.	6.74	2.63	2.28

Table 4.2
Heart Rate from video data of index finger tip vs. Polar T31 monitor (iPhone 4)

READING	HEART RATE (bpm) from Video Data	HEART RATE (bpm) from Polar T31	DIFFERENCE (bpm)
1	88.23	84	4.23
2	78.94	82	3.06
3	78.94	84	5.06
4	88.23	84	4.23
5	78.94	84	5.06
6	88.23	86	2.23
MEAN	83.58	84	3.97
STD. DEV.	5.08	1.26	1.12

From the data presented above, it can be seen that the approach used for heart rate calculation produces results that closely match those of the Polar T31 monitor for both the HTC Desire and the iPhone 4, with the latter being closer (SD = ± 1.12) to the Polar T31's results than the former (SD = ± 2.28). The inability of the iPhone 3GS to provide sufficiently usable data, coupled with its low resolution camera and lack of flash indicates that it is not quite suitable for the methods being used. It was also discovered that the default recording software on the HTC Desire was unable to record audio from an external microphone input alongside the video. This fact, coupled with the fact that writing custom mobile software was out of scope for this study, renders the HTC Desire unsuitable, as the proposed method requires simultaneous audio and video recording capabilities for PWV measurement. Hence, the iPhone 4 has been chosen out of the three possible mobile phones.

<u>Conclusion</u>

While the iPhone 3GS is not suitable for the methods being used here, both the HTC Desire and the iPhone 4 proved to be adequate for good video acquisition. However, due to the recording limitations of the default HTC Desire software mentioned above, the iPhone 4 emerges as the chosen device for all further experiments.

4.1.2   Experiment 2: Determining the most suitable Region of Interest

With the iPhone 4 successfully chosen for further data acquisition, the next step is to identify the particular region on the body that can be used to obtain a clear video signal. This experiment describes the steps taken in examining all the potential regions.

<u>Aim of the Experiment</u>

Four different locations on the body have been chosen as potentially suitable regions for video data acquisition; the forehead, left palm, left index finger tip, and left thumb tip. The aim here is to identify the one region that provides the most suitable results from the acquired video data. Fig. 4.5 visually depicts the regions being examined.
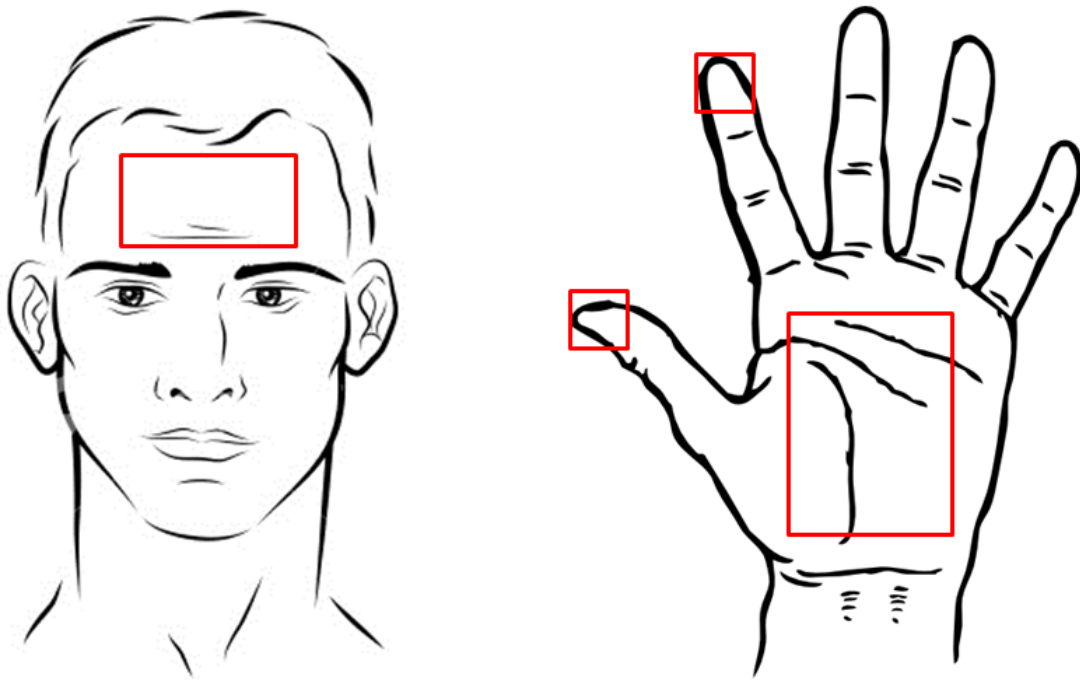
Fig. 4.5 – The four regions to be tested for video acquisition
("Face of a man", n.d.; "Hand 2 clip art", n.d.)

<u>Instruments and Materials</u>

Same as those utilised for Experiment 1 (Section 4.1.1), with the only difference being that the iPhone 4 is the only mobile required.

[57]

<u>Prerequisites</u>

Gel is applied to the Polar T31 monitoring strap and this is fitted around the subject's chest to allow for simultaneous heart rate monitoring.

<u>Setup</u>

The exact setup and recording procedure differs slightly between the chosen regions, and hence the steps for each region are detailed separately.

For the Forehead region:

This procedure requires an additional person to assist with performing the video recording with the phone. When processing the signal from this region, an additional step known as Blind Source Separation was used, based on the work done by Poh et al. (2010) using a freely available MATLAB implementation of the JADE algorithm (Cardoso, 1997). This technique is used to recover unobserved signals from a set of observed mixtures when no information about the mixing process is available (Poh et al., 2010). It results in a single graphical waveform that is based on the three individual colour channels from the original source signal.

1) The subject is seated upright in a chair with a back support to help reduce movements, in a manner that allows them to face the assistant.

2) The mobile phone's camera application is activated and the flash is switched on.

3) The assistant holds and positions the phone in front of the subject's face in a way that allows them to clearly focus on a portion of the forehead. Care must be taken that the subject's hair does not obstruct the region.

4) The assistant begins the video recording on the mobile for a duration of 1 minute. During this time, the subject must breathe normally. At the same time, heart rate monitoring using the Polar T31 is carried out, with results being logged on the PC.

5) Recording is stopped at the 1 minute mark, and the video file is transferred to the PC. Monitoring with the Polar T31 is also stopped.

6) The video file is converted to the AVI format with a frame rate of 25 fps using Xilisoft Video Converter 6.

7) In MATLAB, the AVI file is converted to an uncompressed AVI, and then processed as described in Section 4.1. The resulting signal is then passed through the JADE algorithm to obtain the source-separated signal.

For the Palm region:

As with the forehead, Blind Source Separation was also applied to the signal acquired from this region.

1) The subject is seated with their left arm elevated and folded at the elbow, such that the palm is at heart level. The arm must be supported by a suitable surface so as to reduce strain that may result in unwanted movements.

2) The mobile phone's camera application is activated and the flash is switched on.

3) Using their right hand, the subject holds and positions the phone in a way that allows them to bring a portion of the left palm into focus.

4) The subject begins the video recording on the mobile for a duration of 1 minute. During this time, the subject must breathe normally. At the same time, heart rate monitoring using the Polar T31 is carried out, with results being logged on the PC.

5) Recording is stopped at the 1 minute mark, and the video file is transferred to the PC. Monitoring with the Polar T31 is also stopped.

6) The video file is converted to the AVI format with a frame rate of 25 fps using Xilisoft Video Converter 6.

7) In MATLAB, the AVI file is converted to an uncompressed AVI, and then processed as described in Section 4.1. The resulting signal is then passed through the JADE algorithm to obtain the source-separated signal.

For the Thumb and Index Finger regions:

The procedure is carried out in the exact same manner as described in Section 4.1.1, using the iPhone 4 camera with flash.

For the Palm region:

A section of the processed video signal for the left palm is shown in Fig. 4.6 (a), and the source-separated signal is shown in Fig. 4.6 (b). It is clearly evident that the signal is noisy and lacks distinct peaks in either case, hence making it unsuitable for heart rate calculations using [Eqn. 4.2].

For the Forehead region:

A section of the processed video signal for the forehead is shown in Fig. 4.6 (c), and the source-separated signal is shown in Fig. 4.6 (d). This region was chosen as a potential area based on the work carried out by Poh et al. (2010), wherein they used a segment of the subject's face for analysis. The waveform here is again of poor quality in both cases, which once again makes this location inappropriate for the proposed method. The fact that a second person is required to take the video recording increases the possibility of motion-induced artifacts beyond the subject's control. Moreover, it is a more difficult area to work with, as getting a clear, focused image depends on some extent to the size of the subject's forehead, as well as their hair growth.

For the Thumb and Index Finger regions:

A section of the processed video signals for the left index finger and the left thumb is shown in Fig. 4.6 (e) and Fig. 4.6 (f) respectively. The shape of the waveform for both regions show a similar trend with clear, distinct peaks, and are largely different from

those obtained from the palm and forehead signals. These peaks enable manual calculation of heart rate using [Eqn. 4.2] as described earlier.
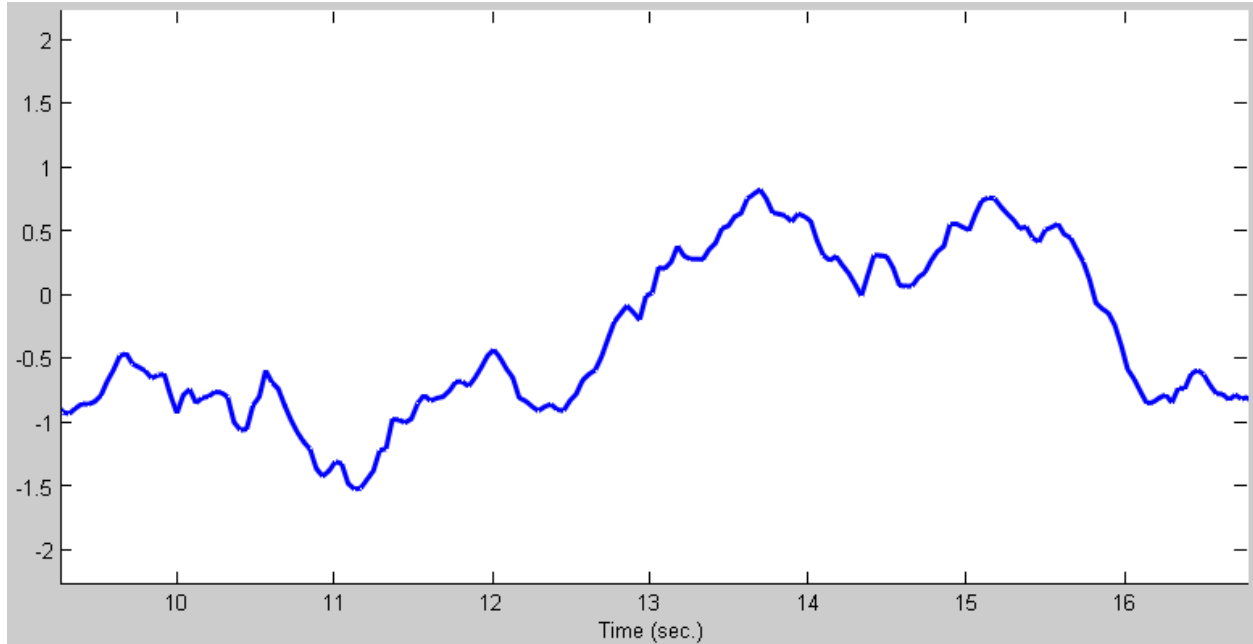
(a) Left palm
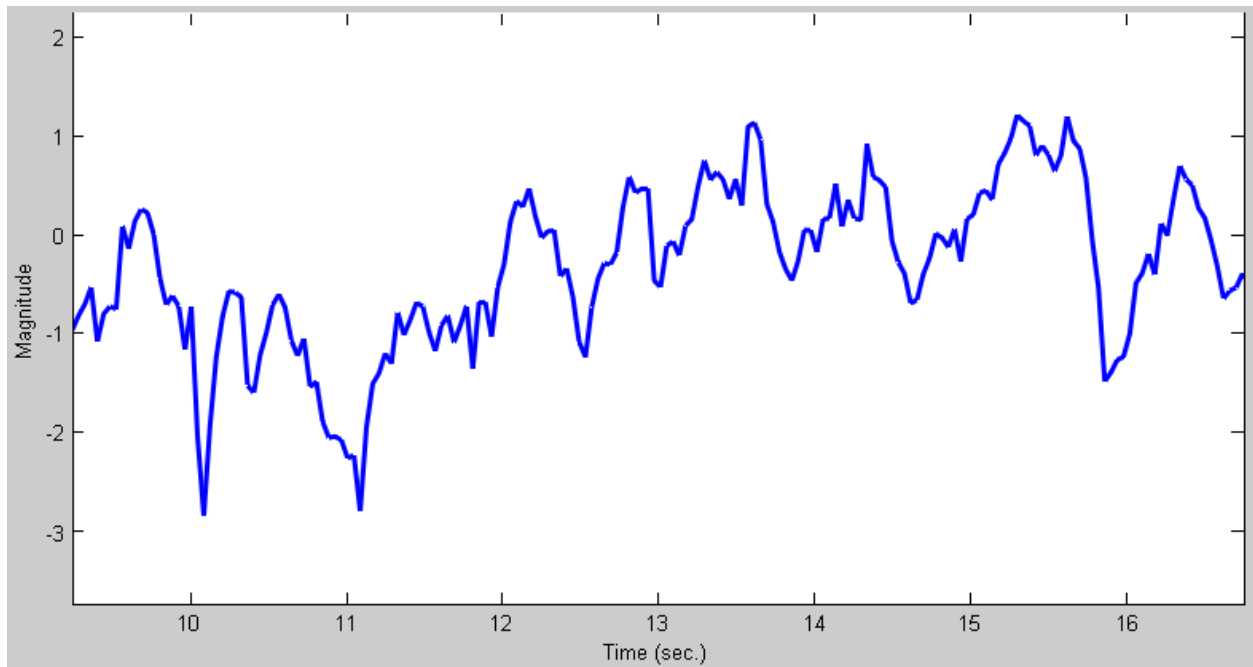
(b) Left palm (source-separated)

[62]
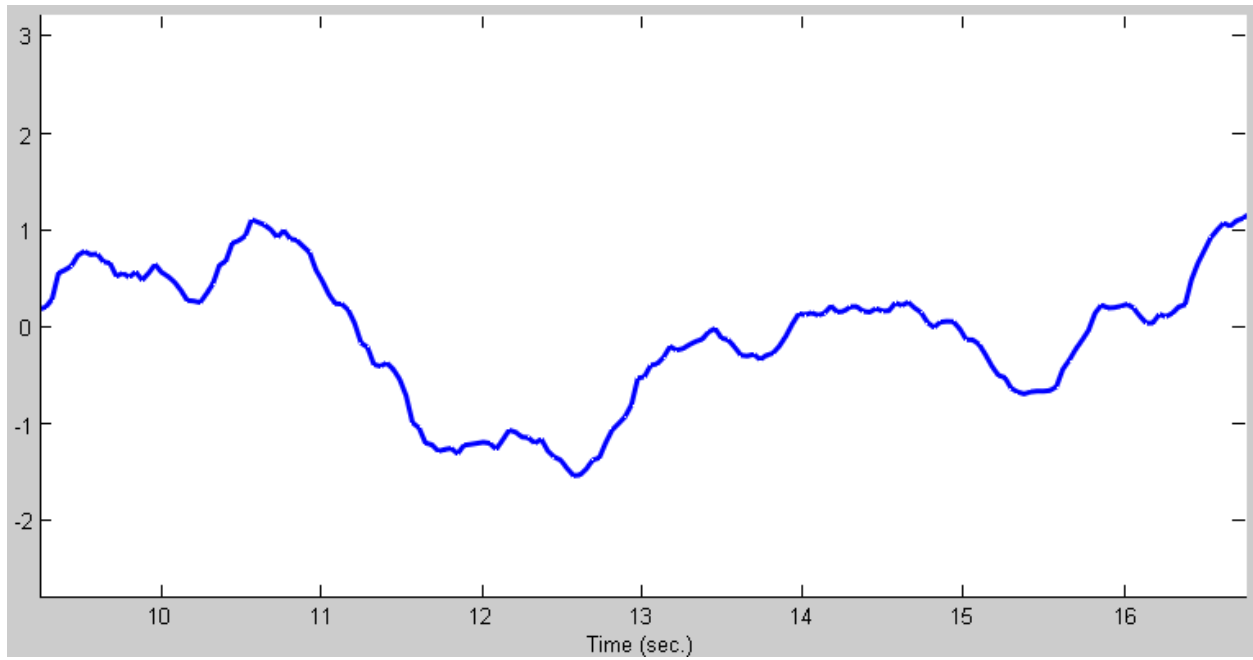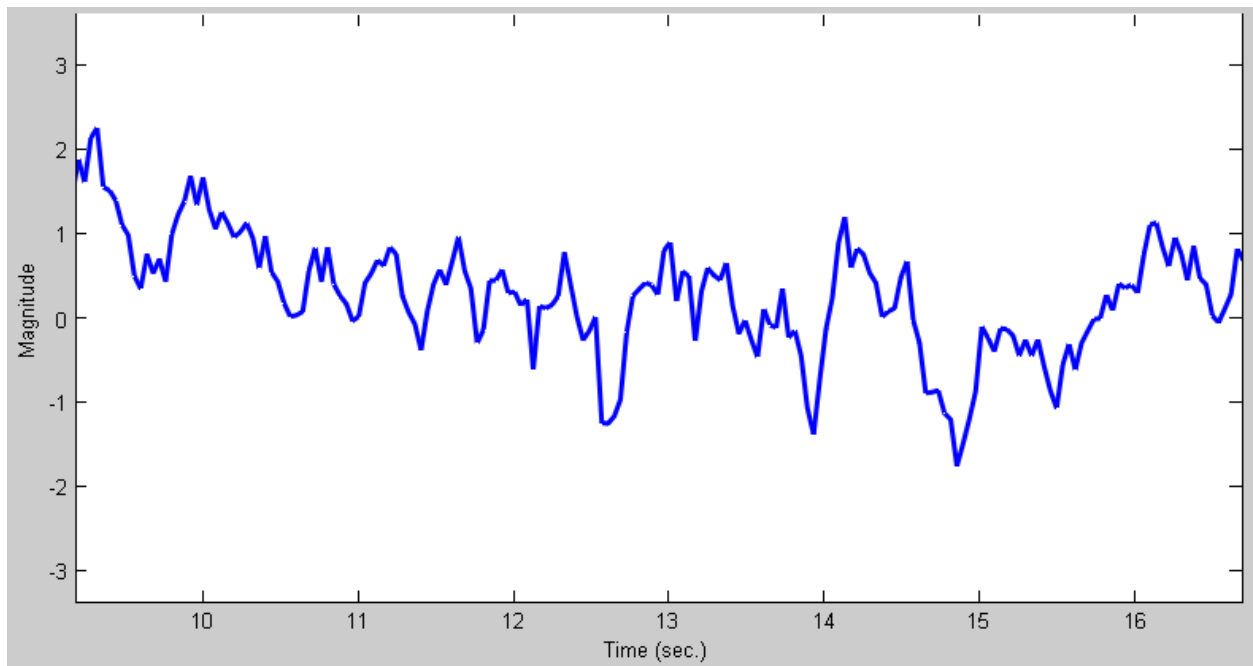
(c) Forehead

(d) Forehead (source-separated)

[63]
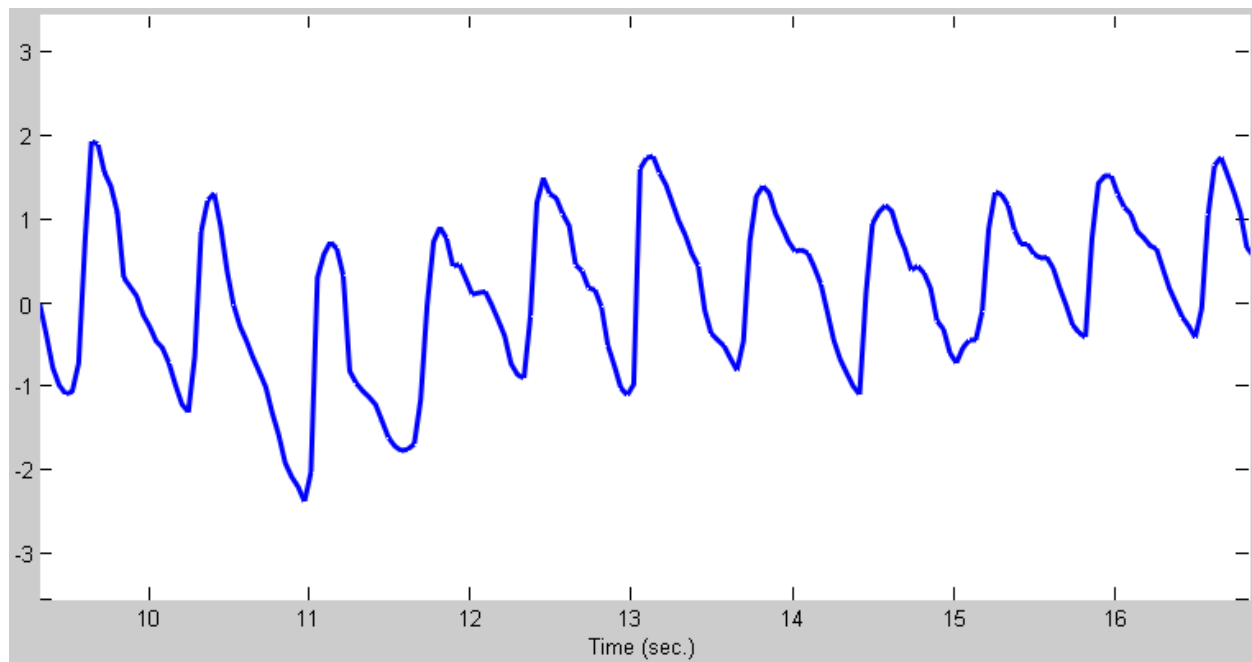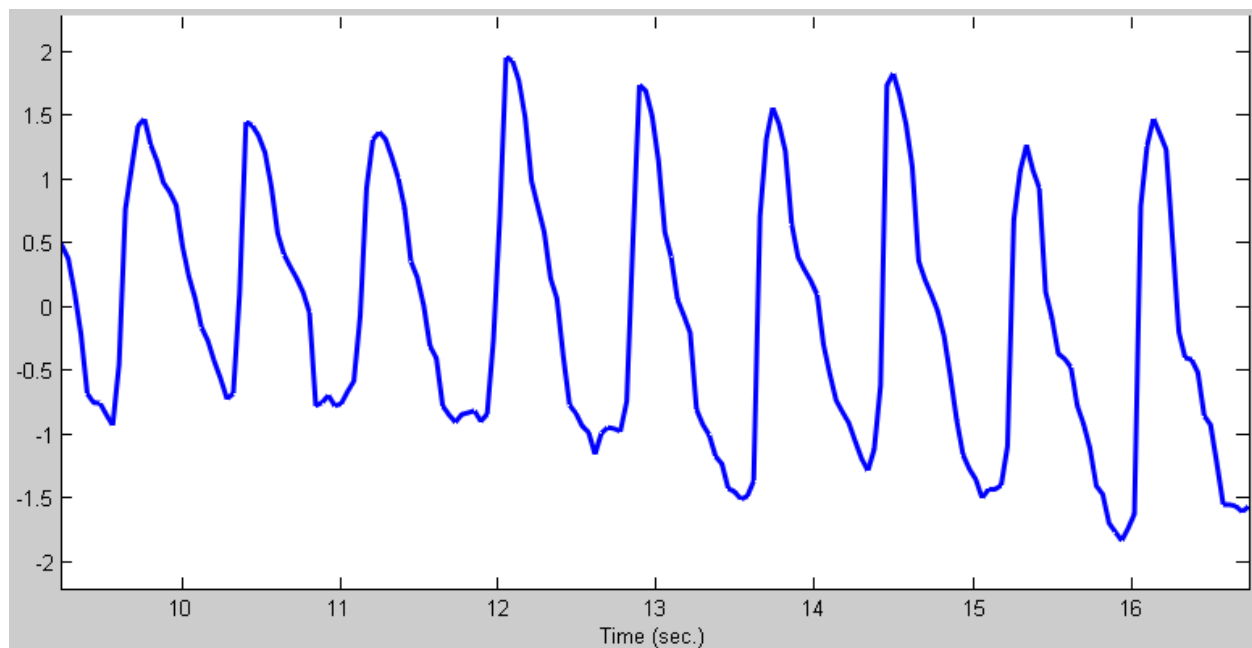
(e) Left Index Finger

(f) Left Thumb

Fig. 4.6 – Section of video waveforms from a recording of each of the chosen locations

[64]

Tables 4.3 and 4.4 present the results of manual heart rate calculation for the thumb and index finger regions, as compared to the Polar T31:

Table 4.3
Heart Rate from video data vs. Polar T31 monitor (Index Finger)

READING	HEART RATE (bpm) from Video Data	HEART RATE (bpm) from Polar T31	DIFFERENCE (bpm)
1	88.23	84	4.23
2	78.94	82	3.06
3	78.94	84	5.06
4	88.23	84	4.23
5	78.94	84	5.06
6	88.23	86	2.23
MEAN	83.58	84	3.97
STD. DEV.	5.08	1.26	1.12

Table 4.4
Heart Rate from video data vs. Polar T31 monitor (Thumb)

READING	HEART RATE (bpm) from Video Data	HEART RATE (bpm) from Polar T31	DIFFERENCE (bpm)
1	83.33	81	2.33
2	83.33	80	3.33
3	75	75	0
4	75	77	2
5	75	72	3
6	71.42	74	2.58
MEAN	77.18	76.5	2.2
STD. DEV.	4.96	3.50	1.17

The figures obtained from the manual calculations are in close proximity to those reported by the Polar T31 for both the index finger and thumb regions, as shown in the tables above. The deviation in results between the thumb (SD = ± 1.17) and index finger (SD = ± 1.12) is marginally different from each other, but the fingertip was chosen due to

[65]

it being the more ergonomic and comfortable option for the end user when performing data acquisition. The index finger site allows a user to hold and operate the mobile phone with the same hand, which is essential when it comes to PWV measurement, due to the introduction of a second handheld device (Doppler probe).

Conclusion

The tip of the index finger and thumb were identified as regions that provided a clear video signal and successful heart rate results that were a close match to those obtained from the Polar T31 monitor. The forehead and palm regions are now disregarded as they have proved to be completely unsuitable for good data acquisition.

The index finger was chosen as the final region of interest to be used for all further data collection for ergonomic purposes, as it will facilitate easier use of the entire setup for PWV measurement as compared to the thumb.

4.1.3 Experiment 3: Validating usability of chosen Mobile Phone and Region of Interest

Aim of the Experiment

The aim here is to ensure that the chosen mobile device and region of interest are capable of producing good results repeatedly and consistently. For this, video data was collected from a subject after vigorous exercise that raised their heart rate considerably, to see if the manually calculated results could still keep up with those from the Polar T31 monitor.

Instruments and Materials

Same as those utilised for Experiment 1 (Section 4.1.1), with the only difference being that the iPhone 4 is the only mobile required.

Prerequisites

Gel is applied to the Polar T31 monitoring strap and this is fitted around the subject's chest to allow for simultaneous heart rate monitoring.

Setup

1) The subject performs some vigorous exercise for a period of 10 minutes to deliberately raise their heart rate. For this case, the exercise was a fast-paced jog on the spot.

2) As soon as the 10 minute period has passed, the subject takes a seated position with their left arm extended forward and elevated to heart level. The arm must be supported by a suitable surface so as to reduce strain that may possibly result in unwanted movements.

3) The mobile phone's camera application is activated and the flash is switched on.

4) The subject holds and positions the phone in a way that the tip of the left index finger covers the camera lens and flash adequately.

5) The subject begins the video recording on the mobile for a duration of 1 minute. During this time, the subject must breathe normally. At the same time, heart rate monitoring using the Polar T31 is carried out, with results being logged on the PC.

6) Recording is stopped at the 1 minute mark, and the video file is transferred to the PC. Monitoring with the Polar T31 is also stopped.

7) The video file is converted to the AVI format with a frame rate of 25 fps using Xilisoft Video Converter 6.

8) In MATLAB, the AVI file is converted to an uncompressed AVI, and then processed as described in Section 4.1.

This procedure was repeated twice more for a total of 3 sets of video recordings and heart rate readings.

Results

A section of the processed video signals for all 3 sets of recordings is shown in Fig. 4.7. The waveforms obtained from processing the signals are similar in shape to those obtained when the subject was at rest, as seen in Experiment 4.1.2.

(a) Set 1

(b) Set 2

[69]
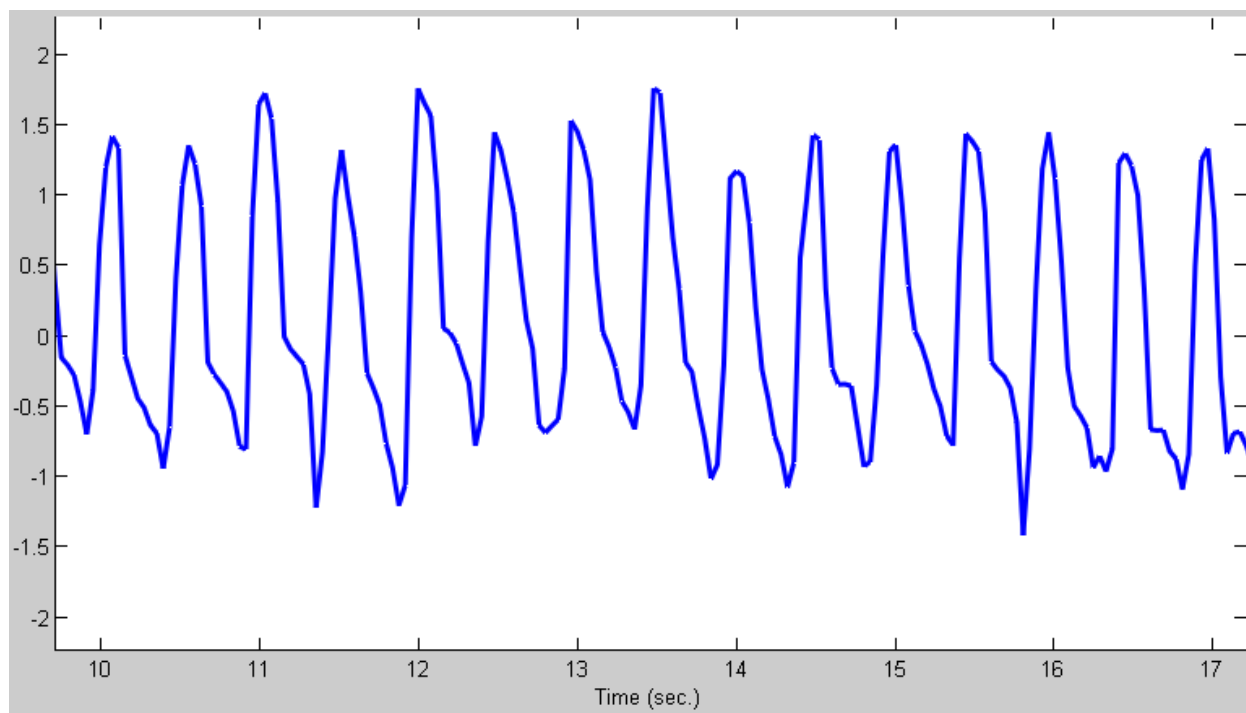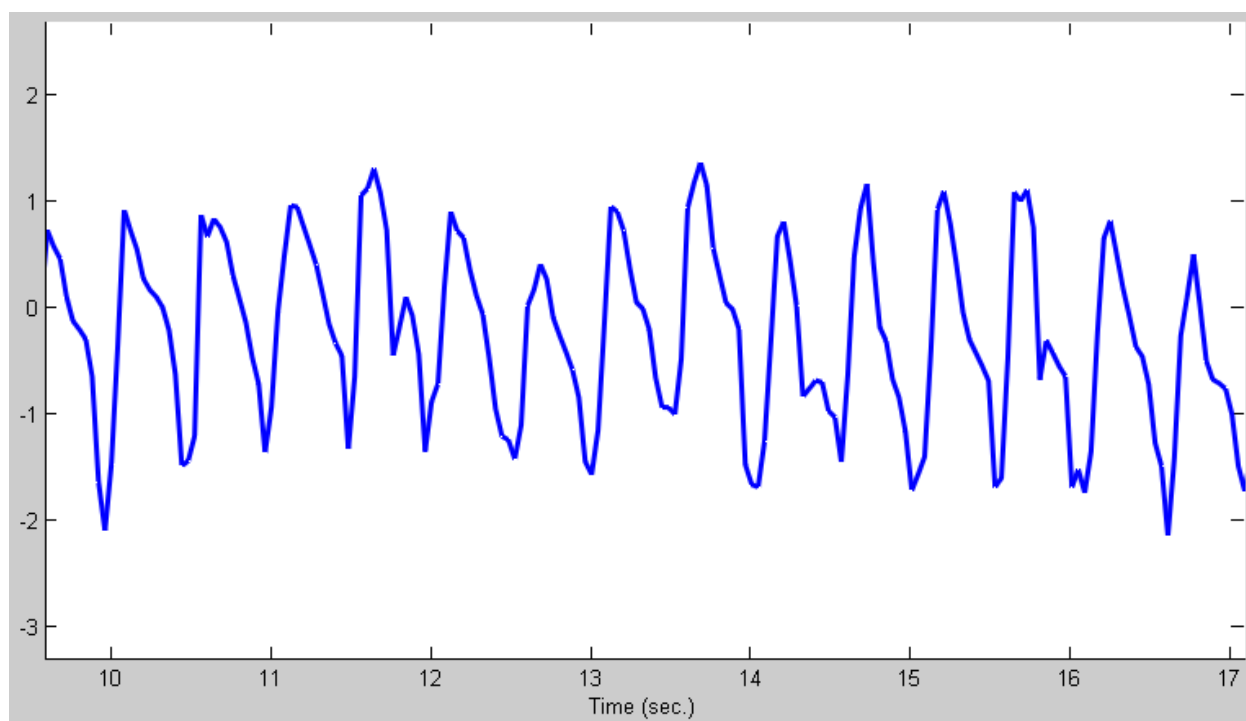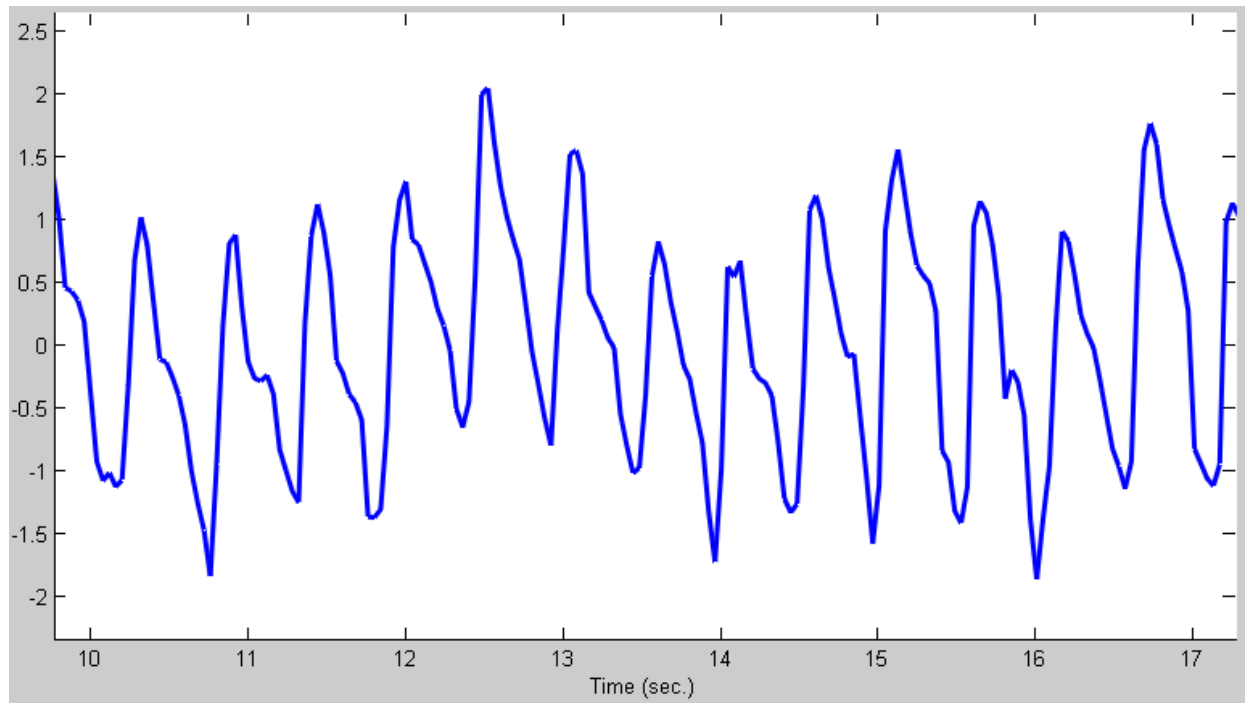
(c) Set 3

Fig. 4.7 – Section of video waveforms from 3 sets of recordings after vigorous exercise

Heart rate was calculated manually in the same way as before, and Table 4.5 presents these results alongside the Polar T31 measurements:

Table 4.5
Heart Rate from video data vs. Polar T31 monitor (10 min. exercise)

SET	READING	HEART RATE (bpm) from Video Data	HEART RATE (bpm) from Polar T31	DIFFERENCE (bpm)
	1	125	129	4
	2	115.38	119	3.62
	3	115.38	118	2.62
1	4	115.38	112	3.38
	5	115.38	113	2.38
	6	115.38	115	0.38
	MEAN	116.98	117.66	2.73
	STD. DEV.	3.92	6.18	1.3

[70]

	1	125	124	1
	2	115.38	118	2.62
	3	115.38	119	3.62
2	4	107.14	115	7.86
	5	125	116	9
	6	125	112	13
	MEAN	**118.81**	**117.33**	**6.18**
	STD. DEV.	**7.41**	**4.08**	**4.54**
	1	100	106	6
	2	107.14	111	3.86
	3	107.14	111	3.86
3	4	107.14	108	0.86
	5	113.20	109	4.2
	6	115.38	112	3.38
	MEAN	**108.33**	**109.5**	**3.69**
	STD. DEV.	**5.42**	**2.25**	**1.65**

As with Experiment 4.1.2, the figures obtained from the manual calculations closely follow the ones reported by the Polar T31 monitor in all 3 sets of recordings.

Conclusion

The results obtained from the video signals using the proposed method of acquisition are consistent and within reasonable range of the reference values from the Polar T31 device, even for much higher heart rates. Hence it has been verified that the combination of the chosen device, region of interest, and measurement method is a valid one, capable of producing comparable results repeatedly.

4.2   Audio Data Acquisition

The audio data is to be acquired by capturing a recording of a subject's heartbeat using the portable Doppler ultrasound probe in conjunction with the mobile phone. As with the video data, a good quality signal is important for appropriate results, and this depends on how clear the acquired audio signal is. To ensure the best possible audio capture, a pair of headphones is also connected to the Doppler probe, as this allows identification of the exact point on the chest from which the clearest sounding heartbeats may be recorded each time.

Compared to the video acquisition process, there are fewer factors that could affect the outcome of the audio signal and thereby influence the results negatively. Positioning of the ultrasound probe and the volume it is set to record at are the main determinants. As with video, noise or distortion can be introduced by unwanted movement of the probe once recording has begun. Out of these factors, the most important one happens to be that of the recording volume setting on the Doppler probe. A very high setting overloads the mobile phone audio input and produces a highly noisy signal which results in clipped peaks and the inability to determine where the actual peaks lie. Conversely, using too low a volume runs the risk of not capturing the audio at all, or producing an irregular signal. In either case, the acquired data would be unsuitable for any further processing and analysis. Hence, a suitable volume must be set to ensure correct audio acquisition, and this too can be determined with the aid of headphones. Fig. 4.8 demonstrates waveforms from audio signals that are normal (a) and highly noisy (b).
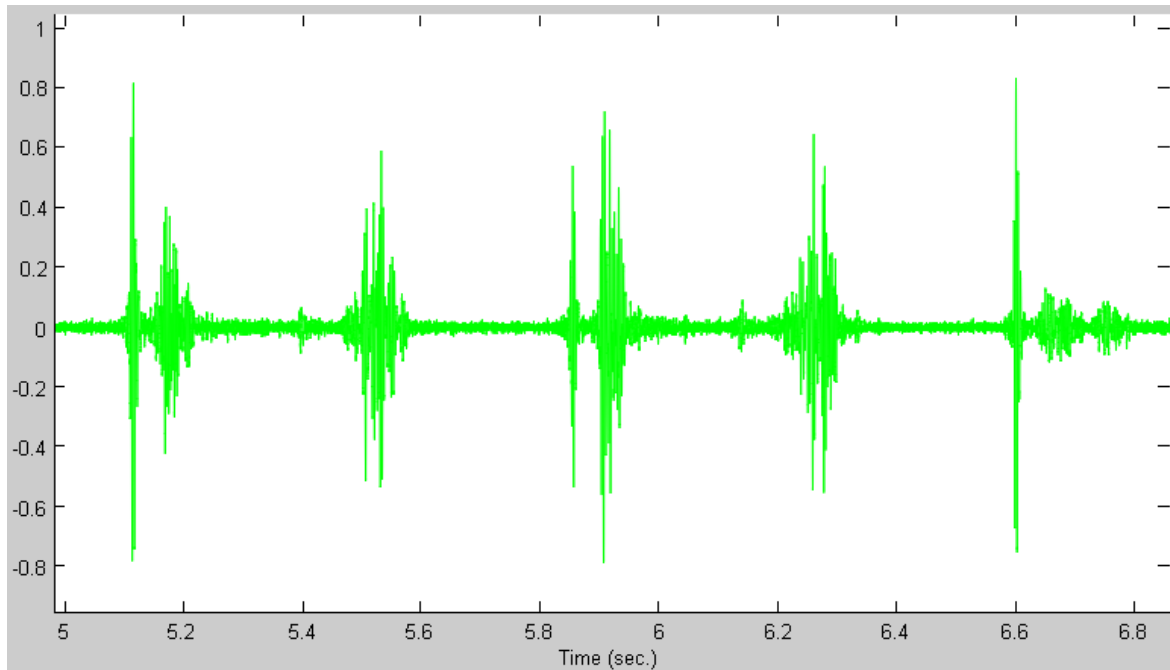
(a) A clean signal at medium recording volume
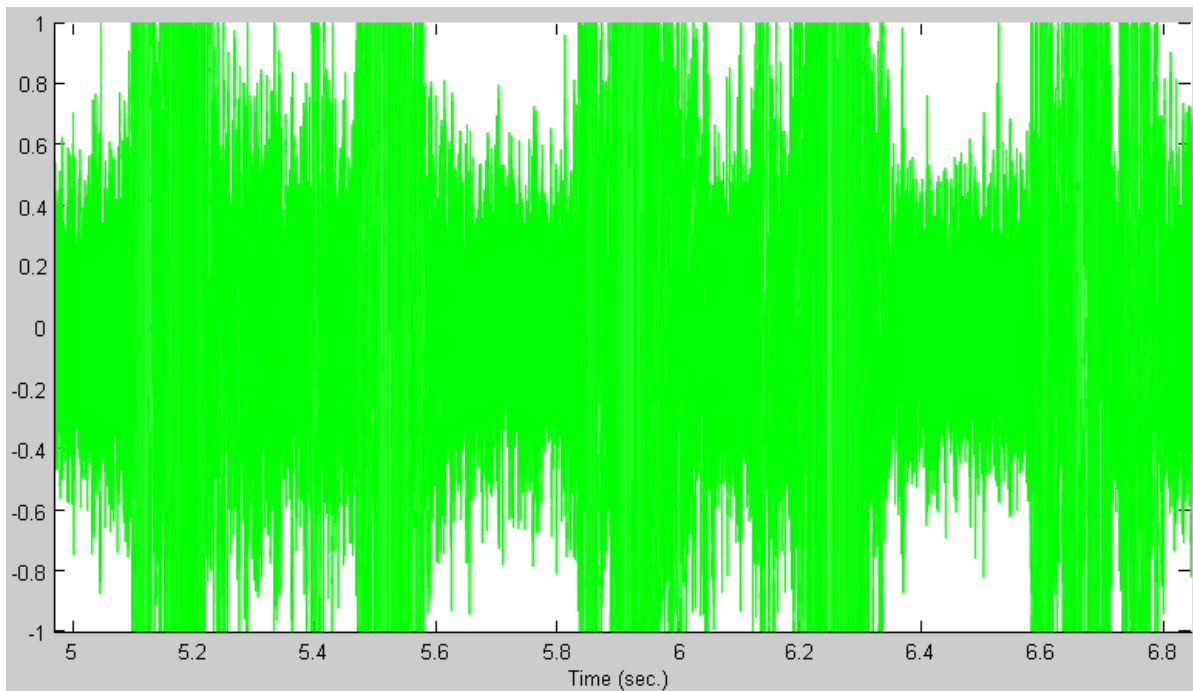
(b) A noisy and distorted signal at high recording volume

Fig. 4.8 – Examples of waveforms produced from audio recordings at different volume levels

To process and inspect the audio signals for heart rate calculation in MATLAB, an approach similar to the one used for video processing was adopted, with the addition of a few more steps unique to the audio signal. The recorded audio file from the phone was first converted to the WAV audio format on the PC to work with MATLAB. The signal was then processed in 30 second blocks of data (as done for the video in Section 4.1), by first passing it through a 6th order Butterworth band-pass filter (100 – 475 Hz), and then extracting the audio envelope by applying a Hilbert transform to the filtered signal. This was done based on work carried out by Lee, Masek, Lam, and Tan (2009), in order to remove noise from the signal and acquire distinct audio peaks. The resulting data was then plotted for manual inspection (Fig. 4.9).
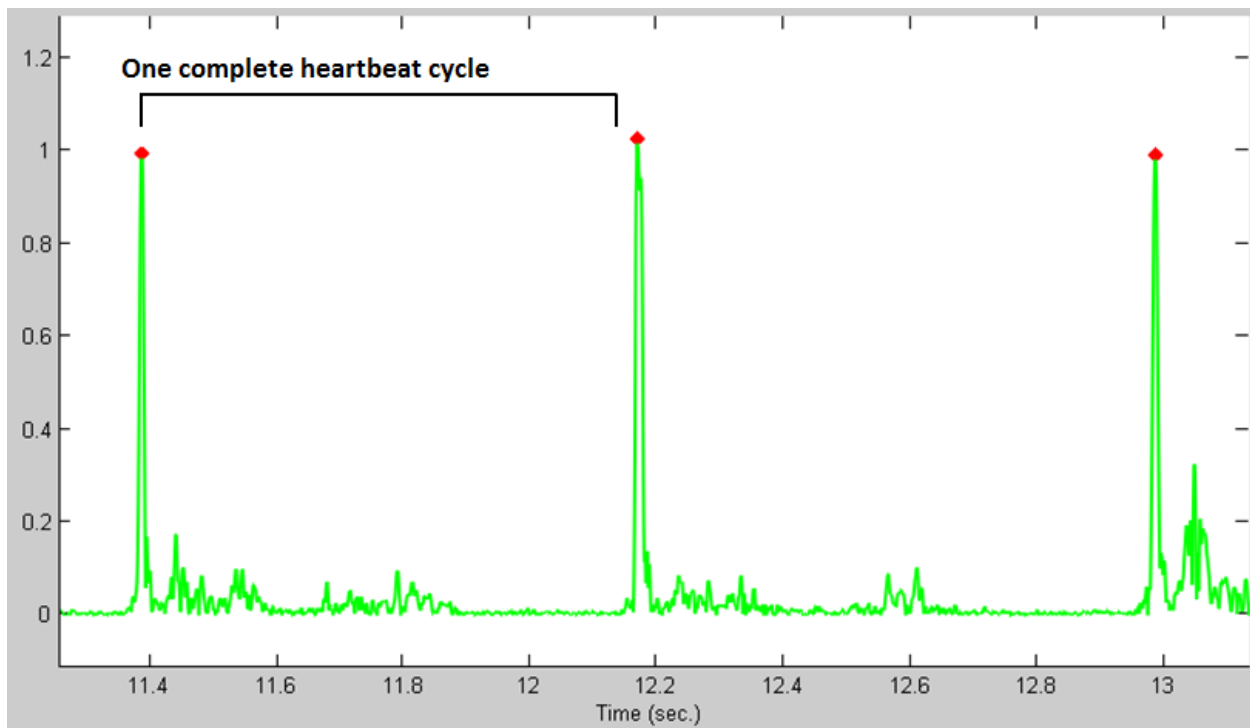
Fig. 4.9 – Audio waveform of a section of a heartbeat recording after being passed through a 6th order Butterworth band-pass filter and Hilbert transform

From the figure, it can be seen that each heartbeat cycle is represented by a set of peaks, one of which is distinctly sharper than the other. Once again, as for the video

[74]

acquisition, measuring the distance between a pair of the sharper peaks (the distance between two heartbeats) allows subsequent computation of the heart rate using [Eqn. 4.2] as described earlier. Fig. 4.9 also illustrates the distance measurement process between consecutive peaks. Again, each data point gives the exact time (in seconds) at which the corresponding peak occurs.

The following presents a complete experiment to carry out the actual process of acquiring audio data via the Doppler probe and mobile phone, and comparing it to results from the Polar T31 monitor to check its validity.

4.2.1   Experiment: Audio signal acquisition via Doppler ultrasound and Mobile Phone

<u>Aim of the Experiment</u>

To investigate the process of acquiring an audio signal from a subject's heartbeat using the Doppler ultrasound probe and chosen mobile phone.

<u>Instruments and Materials</u>

1) Mobile Phone (iPhone 4)

2) Portable Doppler ultrasound probe

3) Stereo cable (3.5 mm to 3.5 mm plug)

4) 1/8" Microphone adapter (3.5 mm 4-conductor TRRS male to 3.5 mm mic input jack)

5) Stereo headphones

6) Windows PC

7) Software
   - MATLAB
   - Xilisoft Video Converter 6

8) Polar T31 heart rate monitor

9) Gel

Prerequisites

Gel is applied to the Polar T31 monitoring strap and this is fitted around the subject's chest to allow for simultaneous heart rate monitoring.

Setup

1) The Doppler probe is connected to the phone with the aid of the Stereo cable and 1/8" microphone adapter plug. The stereo headphones are also connected to the probe.

2) Gel is applied to the tip of the probe to improve the reception of the ultrasound waves.

3) The subject is seated upright in a chair with a back support to help reduce any possible movements.

4) The Doppler probe is positioned against the left side of the chest over the heart area and powered on. The volume switch on the probe is turned up to approximately 25%.

5) Using the headphones, the position of the probe is adjusted until a clear sounding heartbeat is detected. At this step, any required volume adjustments may also be made.

6) The subject begins the audio recording on the mobile for a duration of 1 minute. During this time, the subject must breathe normally. At the same time, heart rate monitoring using the Polar T31 is carried out, with results being logged on the PC.

7) Recording is stopped at the 1 minute mark, and the audio file is transferred to the PC. Monitoring with the Polar T31 is also stopped.

8) The audio file is converted to the WAV format using Xilisoft Video Converter 6.

9) The WAV file is loaded in to MATLAB and then processed.

Results

A section of the processed audio signal recorded from the chest using the Doppler probe and mobile phone is shown in Fig. 4.10. The waveform has good, distinct sets of peaks, indicating good signal acquisition. These peaks then allow for heart rate calculation

using [Eqn. 4.2]. Calculations were performed in the same manner as with video data; a pair of data points was chosen from the waveform at every 10 second interval, thereby giving a total of 6 measurements over the 60 second time period.
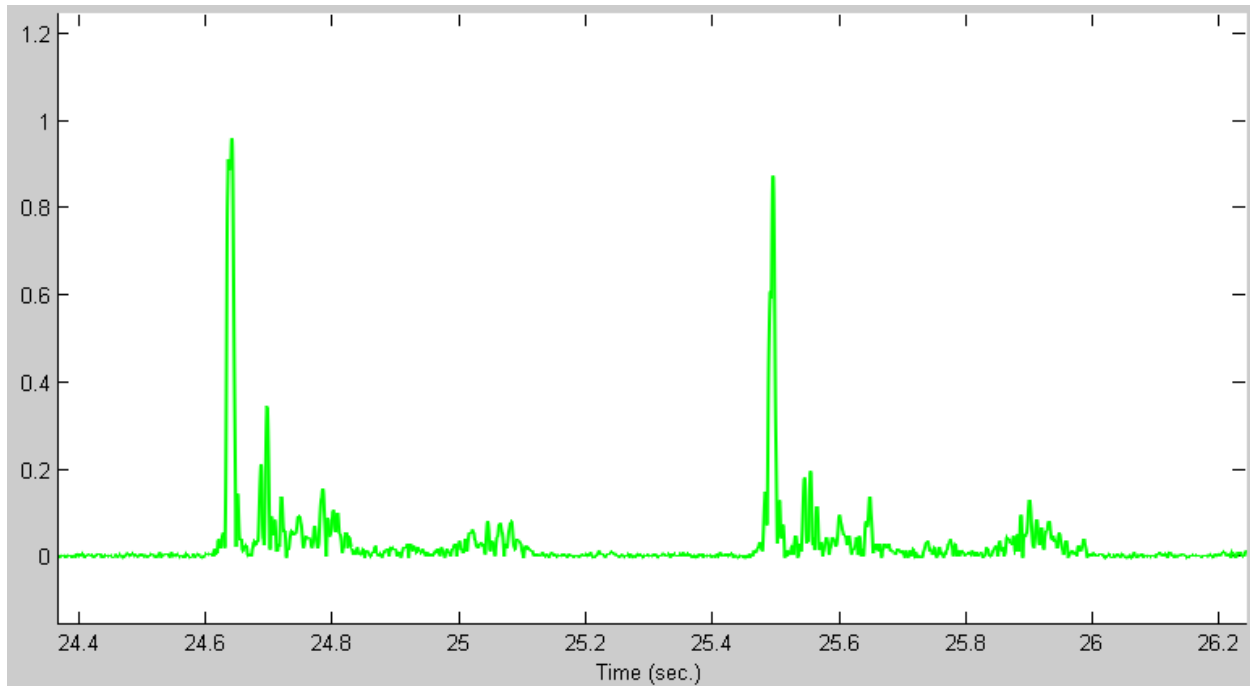
Fig. 4.10 – Section of audio waveform from Doppler probe recording from the chest

The results of the manual calculations as compared to those from the Polar T31 monitor are presented in Table 4.6:

Table 4.6
Heart Rate from audio data vs. Polar T31 monitor

READING	HEART RATE (bpm) from Audio Data	HEART RATE (bpm) from Polar T31	DIFFERENCE (bpm)
1	88.23	84	4.23
2	78.94	82	3.06
3	78.94	84	5.06
4	88.23	84	4.23

5	78.94	84	5.06
6	88.23	86	2.23
MEAN	**83.58**	**84**	**3.97**
STD. DEV.	**5.08**	**1.26**	**1.12**

The figures obtained from manual processing are in close proximity to those reported by the Polar T31 monitor, as shown in the table above.

Conclusion

The waveform obtained from the recorded audio signal and the heart rate computed from it show that the proposed method of audio acquisition using the Doppler probe and mobile phone is capable of producing results similar to the reference values from the Polar T31 monitor, if factors such as Doppler probe placement and recording volume are carefully considered during setup. Hence this method of audio acquisition shall be used for further investigation into measuring PWV.

4.3   Summary

In this chapter, all the important aspects relating to the proposed methods of data acquisition were discussed, including the various factors that needed careful consideration to ensure minimal distortion or skewing of the obtained results. Through a series of experiments, various variables were examined, and the validity of the proposed methods and instruments for data acquisition was demonstrated by computing heart rate manually and comparing it to measurements taken by a commercial heart rate monitor.

Having determined the exact setup for audio and video data acquisition from the experimental results, concrete methods have been established for acquiring data that will allow for PWV calculation based on the two-point measurement technique. The succeeding chapters deal with the various aspects pertaining to PWV measurement using the proposed approach.

5.0   PULSE WAVE VELOCITY ESTIMATION

This chapter deals entirely with the core aspect of the study, which is the computation of PWV from the acquired audio and video signals that have been discussed in Chapter 4. The proposed approach, accompanying equations, and the complete algorithm are all presented here in detail. Experiments that test the usability of the new approach are also provided, and individual results are discussed before the work is brought to a general conclusion.

5.1   Approach and Algorithm

Several different methods exist for acquiring the necessary data to compute PWV, both invasively and non-invasively. Of the non-invasive techniques, the most common ones cited from relevant literature (Section 2.2) include ECG, PPG, applanation tonometry, and Doppler ultrasound. Each of these techniques captures different parameters, and different approaches exist for processing the captured data to get meaningful results.

The basic principle underlying the approach used in this study is similar to that of a two-point measurement via ECG and PPG methods, as discussed in Section 4.0; the collected audio and video data can be likened to an ECG and PPG recording, respectively. As seen through the experiments in Chapter 4, processing the audio and video data yields a waveform for each, and by examining the time difference between two corresponding parts of the signals, PWV can be determined using the following equation:

$$PWV = L \: / \: PTT$$ [Eqn. 5.1]

[81]

where *L* is the distance between the two measurement points, and *PTT* refers to the transit time of the pulse wave, i.e. the time delay between the audio and video signals. Fig. 4.5 illustrates the audio and video signal plots for the purpose of PWV calculation.
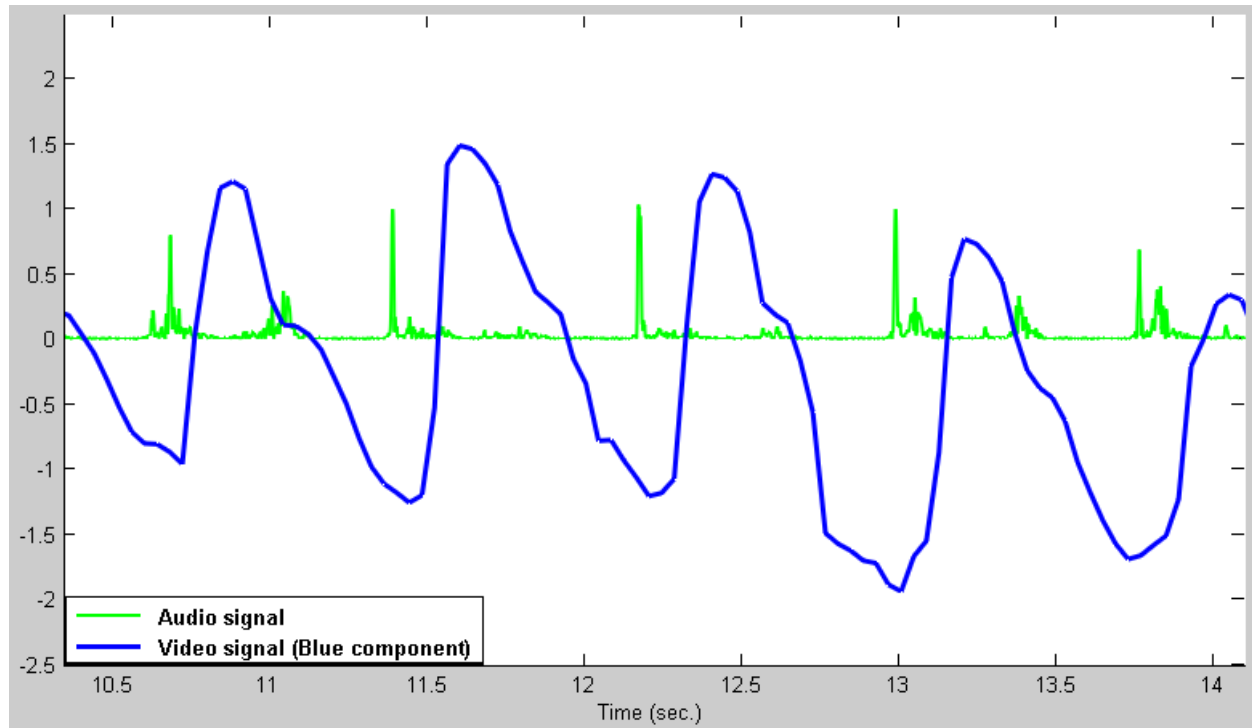
Fig. 5.1 – Section of processed audio and video signals for PWV measurement

While the above equation is a simple and widely-used method to calculate PWV, there are two important factors that need to be discussed regarding its usage:

1) *Measurement of distance between the two recording sites:* This procedure differs based on the regions involved, but the most common method is to make use of a simple tape measure to ascertain the length (in metres) of the arterial segment under investigation (Wilkinson et al., 2010; Hermeling et al., 2007). For the purpose of the proposed solution, the two measurement points are the left side of the chest and the tip of the left index finger. While the tip of the index finger is a fixed and

easily identifiable location, there is no real fixed point on the chest; the placement of the Doppler probe is governed by the availability of a clear signal, and this point may differ from person to person. Therefore, to provide a consistent point for the distance measurement, the sternal notch (the v-shaped depression formed by the collar bones at the base of the neck) is used as the reference point for the chest (Fig 5.2). This location makes sense due to its position with respect to the heart, and the fact that blood pumped from the heart flows outwards and upwards first, thereby passing by the sternal notch on its way down the arm. It has also been used for the same purpose in other studies (Liu et al., 2010; Jiang et al., 2008).

Fig. 5.2 – Location of the sternal notch ("V") on the human body
("The Cardiovascular System", n.d., Carotid artery anatomy)

2) *Measurement of the transit time between signals:* The PTT refers to the amount of time taken (in seconds) by the pulse wave to reach one specific point along the arterial tree since its propagation from some other point, the two points being a known distance apart. To acquire this information from the waveforms, there are two commonly used methods mentioned in the literature: peak-to-peak time calculation (Salvi et al., 2008) or peak-to-foot time calculation (Liu et al., 2010). The

former measures transit time as the difference between two corresponding peaks in the waveforms, whereas the latter measures transit time as the difference between the foot of one waveform and the peak of the other. In the case of PPG and ECG signals, the peak-to-foot method is commonly employed, and since the methods adopted in this study operate on similar principles, the peak-to-foot method shall be used for transit time calculation. Fig. 5.3 illustrates this.
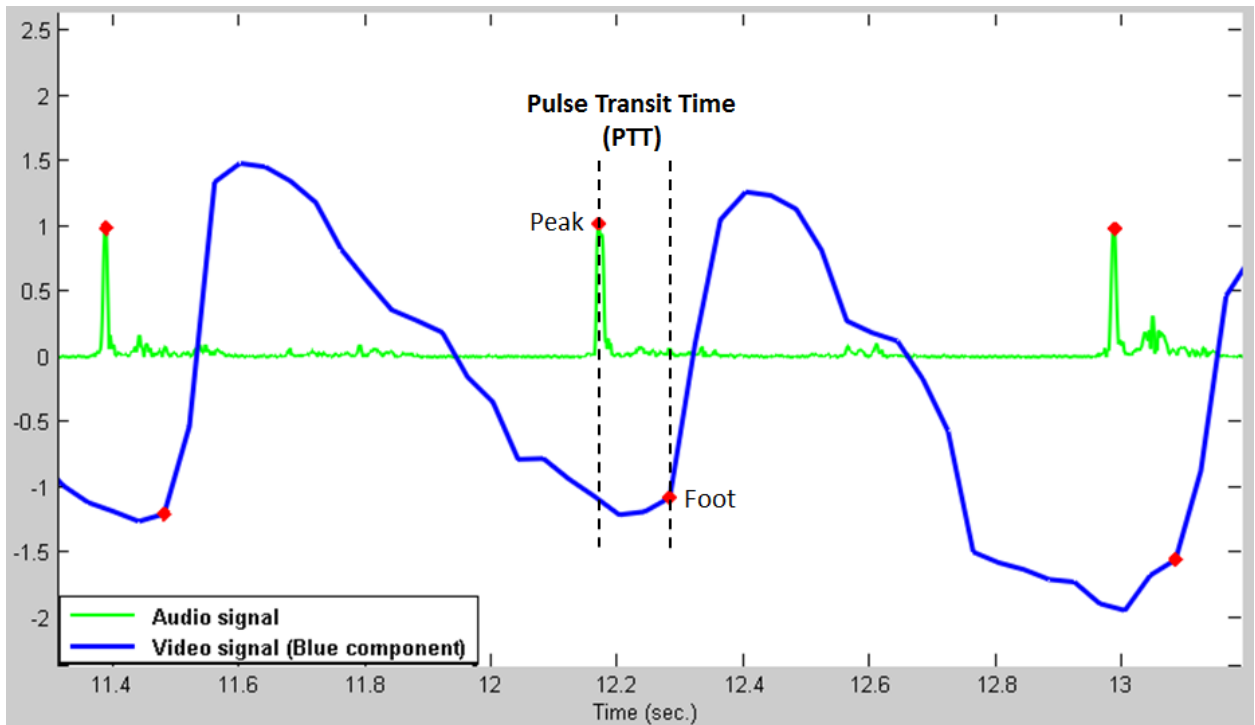
Fig. 5.3 – Peak-to-Foot measurement technique for PTT

Based on the above information, a custom MATLAB program was developed for automatic calculation of PWV from a given audio and video signal. This program made use of the basic data acquisition functionality developed earlier, and extended it by adding in additional steps needed to compute the velocity. The program takes 3 inputs to work with – a video file, an audio file, and the length of the arterial segment being measured (in metres). Processing is then carried out, and the results are calculated and

displayed at the end. The following sub-sections break down the program into the individual processes that make up its total functionality, and discuss each one.

5.1.1   Audio and Video Pre-processing

The pre-processing step refers to the procedures described in Section 4.1 and 4.2 to generate and plot the audio and video waveforms. To briefly recap, this essentially involves the following:

1) *For the video data:* A 30 second sliding window with 1 second increments is used to extract and normalise the red, green and blue channels for all the video frames from the uncompressed AVI file. From this, the blue channel is plotted and used in further processing.

2) *For the audio data:* The same 30 second sliding window as above is used to extract the raw audio data from the given WAV file. This data is passed through a 6th order Butterworth band-pass filter in the range of 100 – 475 Hz to eliminate noise, and the resulting signal is then run through a Hilbert transform to extract the audio envelope. This is the final product that is then plotted and used in further processing.

5.1.2   Peak and Foot Detection of Waveforms

Having obtained waveforms of the processed video and audio data, the program requires a way to be able to estimate the location of a peak in the audio waveform and a foot in the video waveform, as this is needed for the transit time calculation in the

[85]

next stage. For this, the program made use of Peakdet (Billauer, 2011), a freely available MATLAB implementation for peak detection in real-world noisy signals, capable of finding both the maxima and minima. Fig. 5.4 illustrates this on a sample signal.
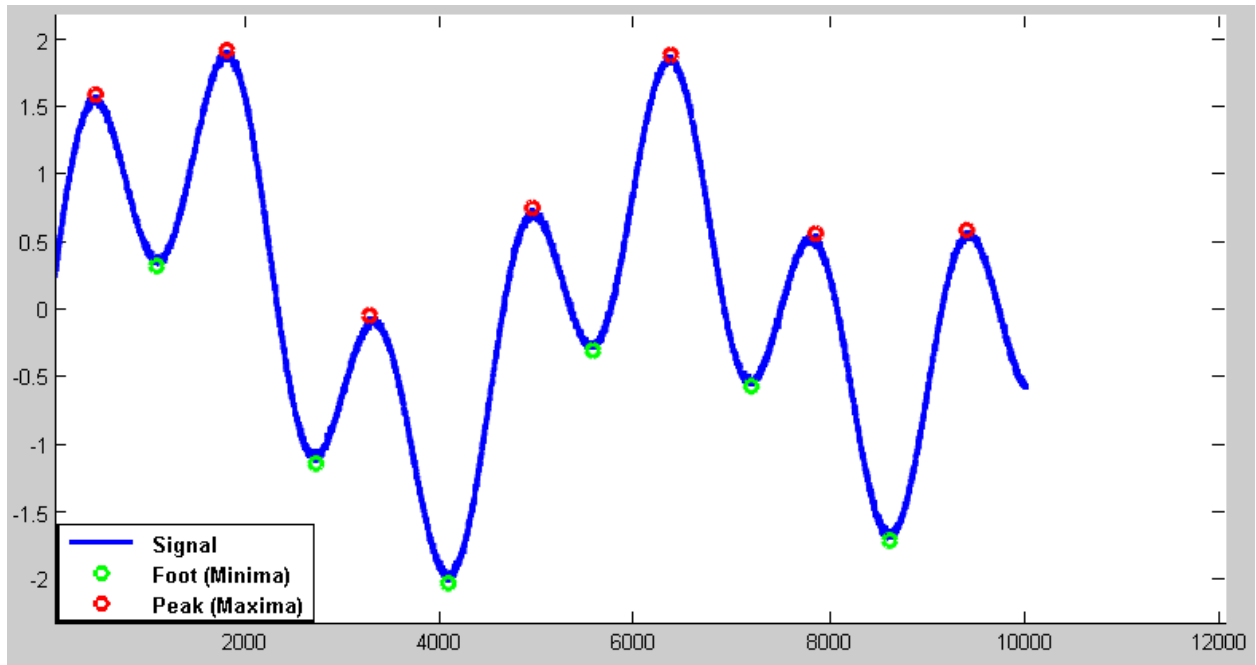
Fig. 5.4 – Sample peak detection in MATLAB using Peakdet

The underlying principle used for peak detection here is the fact that a peak is defined by a point that is higher, by some arbitrary amount, than the other points that immediately surround it. The Peakdet function operates on this principle, taking in a user-specified value to be used as a threshold, and then looping through the signal to look for the highest point around which there exist other points that are lower than the threshold value on both sides (Billauer, 2011). It operates in a similar fashion for the identification of valleys. In this study, threshold values of 0.5 for the video signal and 0.75 for the audio signal were used. Applying the Peakdet function to the processed audio and video signals produced results like those shown in Fig. 5.5. The function returns a set of X-axis values in an array for each signal, each of which represents the

[86]

exact time (in seconds) at which the detected peak or foot occurs. These values are then used to work out the PTT.
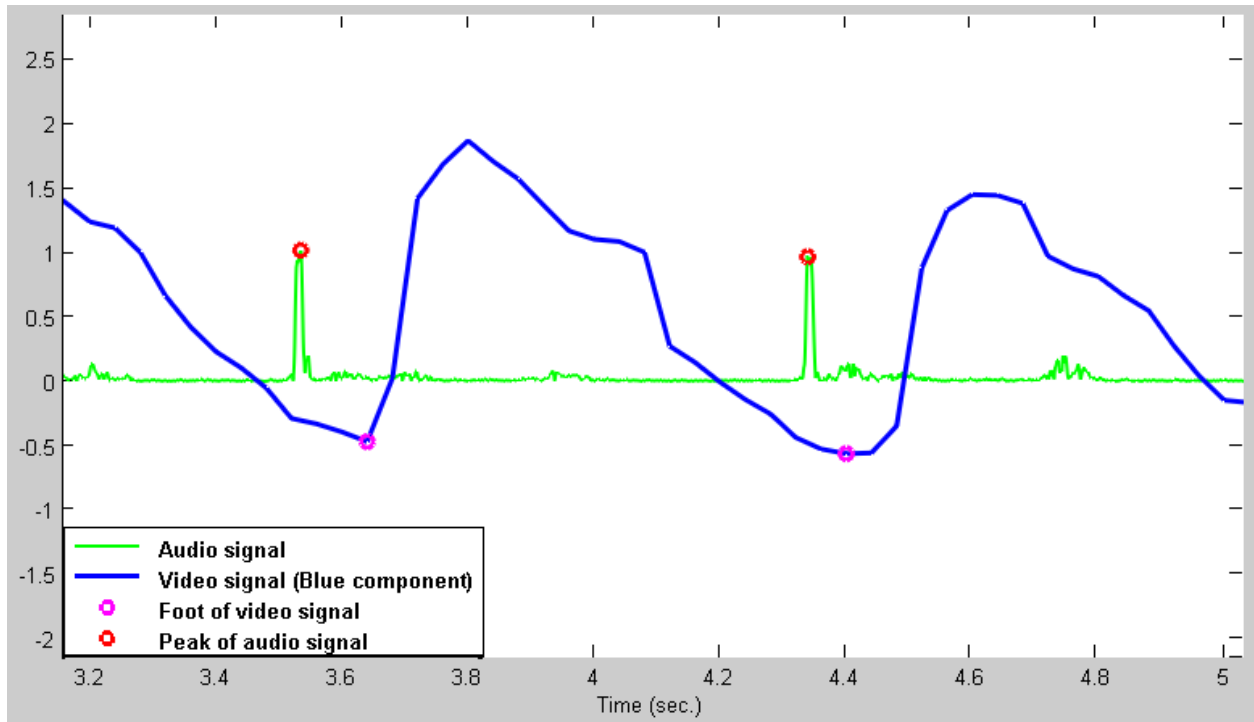
Fig. 5.5 – Results of peak and foot detection on an audio and video signal using Peakdet

5.1.3 Computation of Pulse Wave Velocity

This part of the program performs the core functionality that the proposed solution is based around – the calculation of PWV from the audio and video signals based on the PTT and length of the arterial segment under observation. The following steps are performed by the program to calculate PWV:

[87]

1) For each 30 second window:

a) The first detected audio peak is chosen, and based on this, the corresponding video foot is chosen. This is done by finding the first time value in the array of video feet that is greater than the current audio peak. The process is then repeated for each subsequent audio peak and video foot, to gather a total of 10 sets from the current window.

b) The PWV for each data set obtained above is calculated as the length of the arterial segment divided by the time difference between the audio peak and video foot (as detailed in [Eqn. 5.1]).

c) The values obtained from (b) are checked to determine which ones are valid and which ones are not. Different studies in the field cite different ranges as being valid PWV measurements, as it is affected by the subject's age as well as the arterial segment along which it is measured. Based on the reference values and results mentioned in work carried out by Phillips (2010), Reusz et al. (2010), and Koivistoinen et al. (2007), values in the range of 2 to 15 m/s inclusive were considered valid. Thus at this stage, only values that lie between these limits are saved for further use, while the outliers are discarded.

d) The average of the valid values obtained from (c) is computed and stored. This value is taken to be the average PWV for the current window.

e) The 30 second window is updated to slide forward by 1 second, and steps (a) to (d) are repeated.

2) Once the signals have been completely processed as described above, a number of valid average PWV values from each of the windows are obtained. These values are then averaged again to give the final average PWV of the subject.

While the approach described above is based on known techniques and is capable of producing results, an issue capable of adversely affecting the results was encountered when first implementing the algorithm. This deserves mention before discussing the results obtained, and is presented in the next section.

5.2   Issues Encountered

For PWV calculation using the proposed approach and [Eqn. 5.1], the PTT was calculated using the peak-to-foot method described in Section 5.1. For this method to work however, it is crucial that the waveforms representing each of the signals are synchronised, so that each peak from the audio matches up with the correct foot from the video, and vice versa. If the signals are not in sync, the peak and foot detection phase of the algorithm will end up picking incorrect sets for transit time calculation, ultimately resulting in incorrect PWV.

This is the main issue that was faced with the implementation of the MATLAB program; the algorithm used assumes that both audio and video signals are in sync with respect to each other, but visual inspection of the waveforms revealed that this was not exactly the case. While the signals started out fine and seemed in sync with each other, it was found that eventually, the audio signal would overtake the video signal, resulting in peak and foot mismatches. This situation wherein an audio peak seemed to occur after the corresponding video foot is practically impossible, as the pulse wave is first ejected

from the heart and only then reaches the fingertip, i.e. the audio signal is the first timing reference and the video signal is the second, so the former must always occur before the latter.

The cause of the problem seemed to originate from either the sampling rate of the audio recording or the frame rate of the video recording on the iPhone 4. Examining the properties of the recorded files showed an audio sampling rate of 44.1 kHz and a video frame rate of 29 frames per second. While the former meets the standard specification for CD-quality audio and remains fixed irrespective of any external factors, the latter does not match any of the video recording standards. Furthermore, the mobile phone and recording software itself allows no control over the frame rate; it is automatically adjusted internally. Hence, an investigation into the possibility of the frame rate negatively affecting the video signal was required. This was done through a simple experiment, as detailed below.

5.2.1 Experiment: Video Frame Rate for Audio/Video Signal Synchronisation

<u>Aim of the Experiment</u>

To determine the correct frame rate required for proper synchronisation between the video and audio signals for the purpose of correct peak and foot detection.

<u>Instruments and Materials</u>

1) Mobile Phone (iPhone 4)

2) Portable Doppler ultrasound probe

3) Stereo cable (3.5 mm to 3.5 mm plug)

4) 1/8" Microphone adapter (3.5 mm 4-conductor TRRS male to 3.5 mm mic input jack)

5) Stereo headphones

6) Windows PC

7) Software

- MATLAB
- Xilisoft Video Converter 6

8) Gel

9) Tape measure

<u>Prerequisites</u>

The distance between the two measurement sites (sternal notch and left fingertip) must be measured and recorded.

<u>Setup</u>

1) The Doppler probe is connected to the phone with the aid of the Stereo cable and 1/8" microphone adapter plug. The stereo headphones are also connected to the probe.

2) Gel is applied to the tip of the probe to improve the reception of the ultrasound waves.

3) The subject is seated upright in a chair with a back support to help reduce any possible movements.

4) The Doppler probe is positioned against the left side of the chest over the heart area and powered on. The volume switch on the probe is turned up to approximately 25%.

5) Using the headphones, the position of the probe is adjusted until a clear sounding heartbeat is detected. At this step, any required volume adjustments may also be made.

6) The subject extends their left arm forward, elevated to heart level. The arm must be supported by a suitable surface so as to reduce strain that may result in unwanted movements.

7) The mobile phone's camera application is activated and the flash is switched on.

8) The subject holds and positions the phone in a way that the tip of the left index finger covers the camera lens and flash adequately.

9) The subject begins recording on the mobile for a duration of 1 minute. During this time, the subject must breathe normally.

10) Recording is stopped at the 1 minute mark, and the video file is transferred to the PC.

11) The video file is converted to the AVI format with a frame rate of 29 fps as well as 25 fps (standard PAL frame rate specification) using Xilisoft Video Converter 6.

12) In MATLAB, the video and audio from each file is separated (to uncompressed AVI and WAV formats) and then processed as described in Section 5.1.

Results

Fig. 5.6 shows a section of audio and video waveforms from the recording for both, the 29 fps video file (a) and the 25 fps video file (b). Visual inspection of the 29 fps video signal (Fig. 5.6 – a) clearly demonstrates the synchronisation problem described earlier; the audio signal overtakes the video signal from about five seconds into the video, with each peak occurring after each foot, something that is technically not possible. The situation seems to get worse as time progresses, with larger differences between each peak and foot set. In comparison, the video signal at 25 fps (Fig. 5.6 – b) stays in sync with the audio from start to finish, as a result of which each audio peak occurs before the corresponding video foot. This allows the points to match up correctly as required for the transit time calculation.
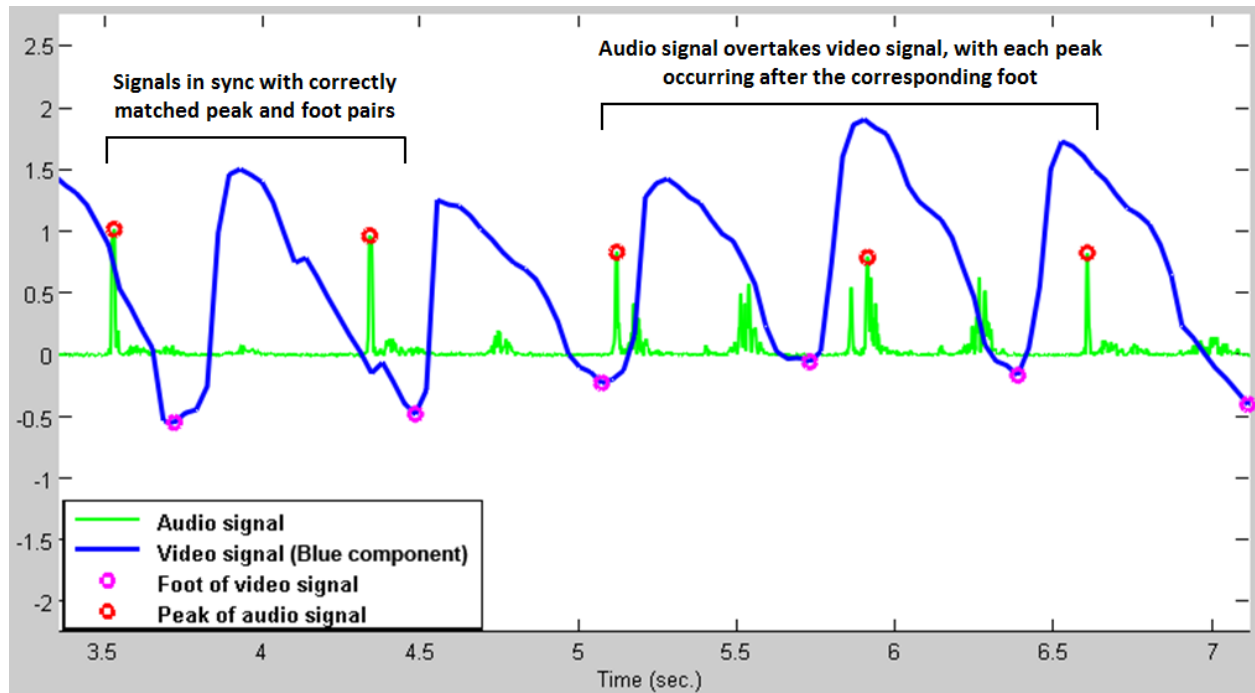
(a) Video at 29 fps
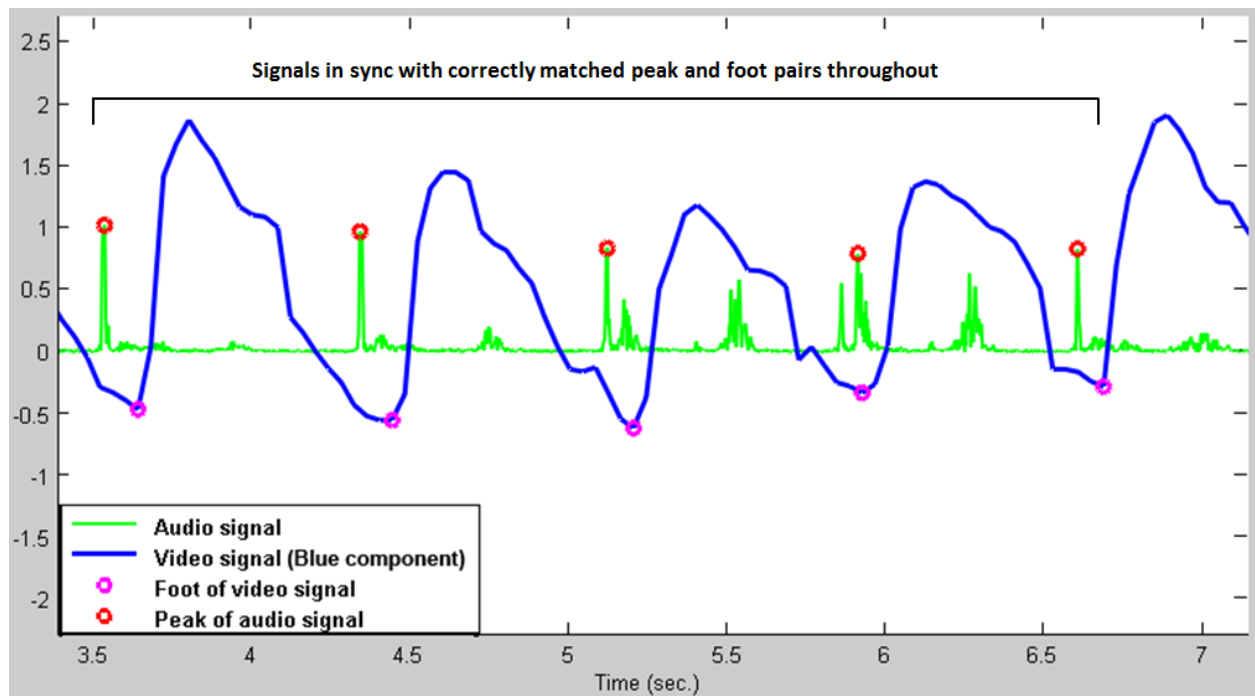
(b) Video at 25 fps

Fig. 5.6 – Audio/video waveforms for PWV with two different video frame rates

<u>Conclusion</u>

By manually converting the video frame rate to a standard 25 fps before processing, the synchronisation issue between the audio and video signals that occurred at the default 29 fps was solved. PTT can thus be calculated correctly, which will in turn allow correct PWV calculations. Hence, it was determined that providing a video file that has a frame rate of exactly 25 fps is essential to the correct working of the algorithm.

Having discussed the main problem encountered with the proposed approach and its subsequent resolution, it is now possible to look at the PWV results obtained from multiple sets of audio and video data.

5.3   Pulse Wave Velocity Experiments and Results

As before, experiments were carried out to acquire audio and video data, for the purpose of calculating PWV using the custom MATLAB program that implements the developed algorithm as described in Section 5.1. Due to the scope and time constraints of this study, the test subjects were limited to members of the research team only. Recordings and measurements were secured from two subjects under a set of different circumstances, and processed to calculate their PWV. The results obtained are discussed with reference to information presented in relevant literature, as the research team did not have access to a commercial PWV monitor. Information about the subjects relevant to these experiments is presented in Table 5.1.

Table 5.1
Information on Subjects involved in PWV measurement

SUBJECT	GENDER	AGE (years)	LENGTH OF ARTERIAL SEGMENT OBSERVED (metres)
1	Male	22	0.99
2	Male	36	0.93

5.3.1 Experiment: PWV measurement under different conditions

Aim of the Experiment

To determine the PWV of subjects under a set of different conditions.

Instruments and Materials

Same as those utilised for the experiment in Section 5.2.1.

Prerequisites

The distance between the two measurement sites (sternal notch and left fingertip) must be measured and recorded.

Setup (A)

To calculate PWV for each subject at rest, the procedure is carried out in the exact same manner as described in Section 5.2.1.

Setup (B)

The following steps were performed once for each subject to calculate PWV after engaging in some vigorous physical activity for a short period of time:

1) The subject performs some vigorous exercise for a period of 10 minutes to deliberately raise their heart rate. For this case, subject 1 performed a fast-paced jog on the spot, and subject 2 made use of an exercise bike.

2) As soon as the 10 minute period has passed, the subject takes a seated position and proceeds to obtain an audio and video recording in the exact same manner as described for when the subject was at rest (Setup A).

Setup (C)

The following steps were performed to calculate PWV in the supine position (lying down with face up) after the subject had rested for some time:

1) The subject rests in the supine position on a bed for a period of 1 hour.

2) At the end of the rest period, the subject is woken and remains lying in the supine position for monitoring.

[97]

3) The Doppler probe is connected to the phone with the aid of the Stereo cable and 1/8" microphone adapter plug. The stereo headphones are also connected to the probe.

4) Gel is applied to the tip of the probe to improve the reception of the ultrasound waves.

5) The Doppler probe is positioned against the left side of the chest over the heart area and powered on. The volume switch on the probe is turned up to approximately 25%.

6) Using the headphones, the position of the probe is adjusted until a clear sounding heartbeat is detected. At this step, any required volume adjustments may also be made.

7) The subject keeps their left arm fully extended by their side, supported by the bed surface so as to reduce strain that may result in unwanted movements.

8) The mobile phone's camera application is activated and the flash is switched on.

9) The subject holds and positions the phone in a way that the tip of the left index finger covers the camera lens and flash adequately.

10) The subject begins recording on the mobile for a duration of 1 minute. During this time, the subject must breathe normally.

11) Recording is stopped at the 1 minute mark, and the file is transferred to the PC.

12) The file is converted to the AVI format with a frame rate of 25 fps using Xilisoft Video Converter 6.

13) In MATLAB, the video and audio from the file is separated (to uncompressed AVI and WAV formats) and then processed as described in Section 5.1.

Results

The acquired data was put through the MATLAB program, and the final PWV was computed by averaging out the average velocities from each 30 second window, as detailed in Section 5.1.3. These results are presented in Table 5.2.

Table 5.2
PWV of two subjects at rest and after 10 min. of exercise

| CONDITION	SUBJECT	AGE (years)	MEAN PULSE WAVE VELOCITY (metres/sec)	STANDARD DEVIATION (of 31 avg. values from 30 sec. windows)
At Rest	1	22	3.295	0.404
	2	36	3.376	1.19
After 10 min. of exercise	1	22	4.457	0.355
	2	36	4.194	0.593

| Supine after 1 hr. of rest	1	22	2.749	0.504

Fig. 5.7 provides examples of audio and video waveforms from a subject at rest (a) and a subject after having completed some vigorous physical activity (b). In the latter case, multiple peaks appear in the audio waveform for each heartbeat segment, as opposed to the former which normally exhibits a single, distinct peak. This behaviour can be attributed to the increased heart rate as a result of the exercise performed, as the heart beats much faster and the subject's respiration is more pronounced. In such cases, the algorithm will end up detecting and using multiple peaks for PWV calculation, which is not desirable. Hence for this case, two additional steps were performed:

1) The processed audio signal was passed through a 5th order Butterworth low-pass filter with a cutoff frequency of 2 Hz (Lee et al., 2009) to smooth out the signal and allow for better, distinct peak formations.

2) To ensure the algorithm would find the smoothened out peaks, the peak threshold for the Peakdet function was changed from 0.75 to 0.1.
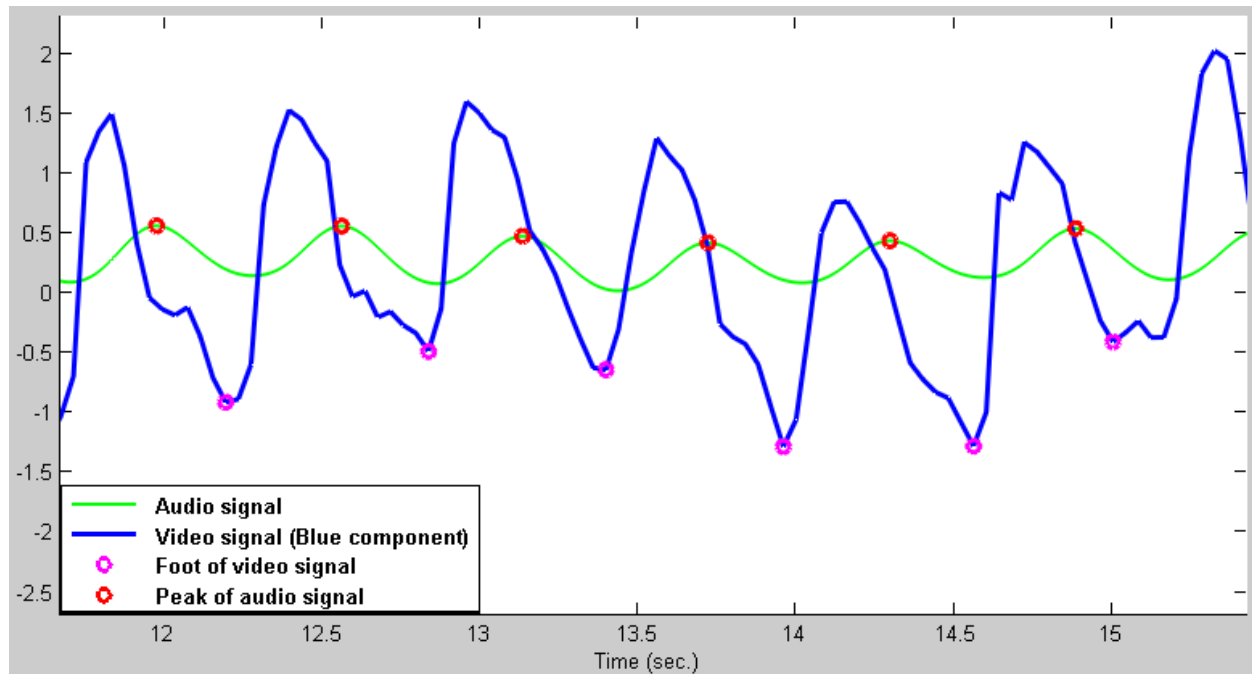
By performing these two steps, the algorithm was able to function as expected, picking out a single audio peak that matched with a corresponding video foot. Fig. 5.7 (c) shows the effect of performing the abovementioned steps on the audio signal.
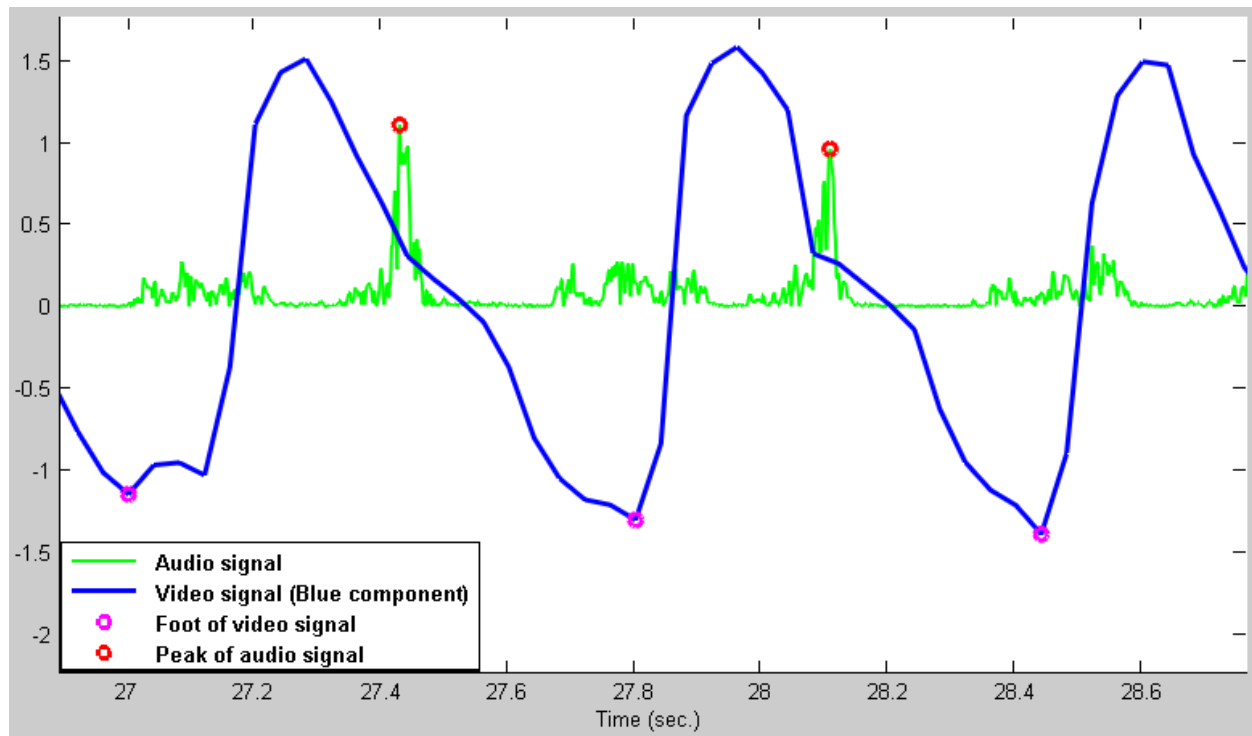
(a) Subject at rest
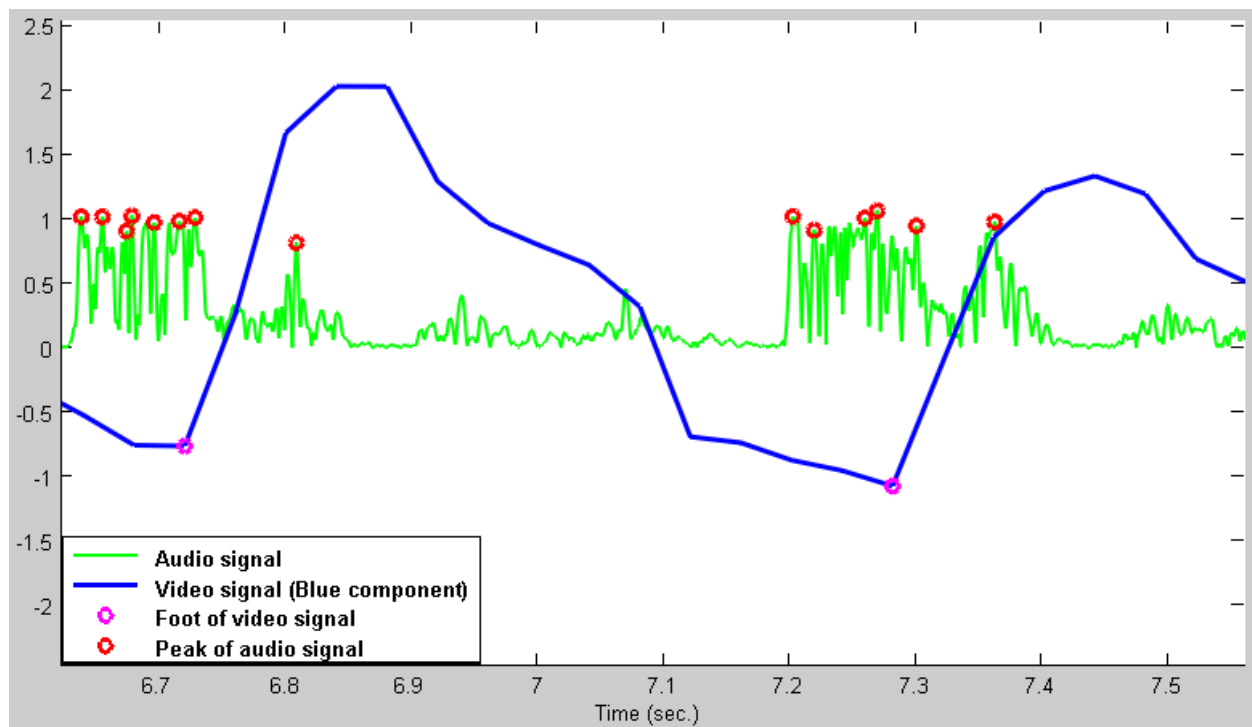
(b) Subject after 10 min. exercise

[101]

(c) Subject after 10 min. exercise with low-pass filtered audio

Fig. 5.7 – Section of audio/video waveforms for PWV measurement experiments

Since a commercial validated PWV monitor (such as Complior or SphygmoCor) was not available during this study, results from the experiments were examined based on relevant literature in the field. Of importance to this study are the following facts:

1) PWV has a positive relationship with age (Korpas et al., 2009), meaning it increases with an increase in age due to the inherent thickening of the arterial walls.

2) PWV has a tendency to gradually increase with an increase in heart rate, as reported by Lantelme et al. (2002).

3) Normal values of PWV range from 2 to 7 m/s in healthy adults (Phillips, 2010), but this value is highly dependent on other physiological factors, and values up to 15 m/s in elderly subjects have been reported (Lantelme et al., 2002).

[102]

Applying these facts to the results obtained from the experiments (Table 5.2), it becomes evident that the method of PWV estimation adopted here exhibits similar trends, as explained below:

1) PWV was the lowest (2.74 m/s) when measured after the subject had rested or slept for an hour and then woken up. This is logically expected due to the fact that the heart rate is much lower when one is asleep as the body is able to relax completely.

2) In both subjects, the velocity after performing exercise is greater than the velocity at rest. This conforms to the claim made by Lantelme et al. (2002) regarding the relation between PWV and heart rate.

3) Both subjects were healthy and free from any chronic ailments that may have influenced the outcome (Eg: hypertension, diabetes), and thus their pulse wave velocities at rest fall well within the normal range for a healthy adult (2 to 7 m/s). Had there been a significantly older subject involved, a much higher number would be expected, depending on the age and health conditions.

Conclusion

Based on the results presented above for the three different conditions under which subjects were tested, it can be asserted that the approach and algorithm adopted provides a fairly simple way to estimate PWV using a mobile phone and Doppler ultrasound. The results are in line with the reference values found in relevant literature and also conform to other standards established by related work in the field.

6.0   DISCUSSION AND FUTURE WORK

PWV is an important physiological phenomenon in the human body. Apart from having been established as the gold standard for arterial stiffness, it shares a relationship with other vital factors such as BP and heart rate. Thus its monitoring can prove useful in certain health conditions. However, commercial equipment for this purpose is expensive and requires special training to operate, and as such, is not suitable for personal use.

Through this research project, a new and unique approach to estimating PWV by means of audio and video data has been devised. The methods presented are derived from similar work using PPG and ECG signals, but to our knowledge, no other study has proposed or made use of the methods and algorithms presented here. The solution described by this study also offers the advantage of using easily available portable devices that are relatively inexpensive when compared to the more clinic-oriented medical grade equipment.

The results obtained from the experiments conducted are in accordance with the reference values published in other relevant work. Experiments were carried out on two healthy male subjects from within the research team, aged 22 (Subject 1) and 36 (Subject 2). The average PWV under normal resting conditions was found to be 3.295 m/s for Subject 1 and 3.376 m/s for Subject 2. This difference may be indicative of the fact that age affects the PWV, but a firm conclusion cannot be drawn based on only two subjects at this time. The magnitude of the difference is small as the age difference between the subjects is not vast; a faster velocity can be expected if there were elderly subjects involved. Velocity of the pulse is also known to rise gradually with an increase in heart rate, and this was also demonstrated by the measurements obtained after the subjects performed some exercises for a ten minute period. Subject 1 performed an on-

the-spot jog and recorded an average velocity of 4.457 m/s, while Subject 2 used an exercise bike and recorded an average velocity of 4.194 m/s. Conversely, PWV is also lower at lower heart rates, as is indicated by the result of 2.749 m/s produced by Subject 1 after resting in the supine position for one hour. This is due to the fact that heart rate is lowest when the body is resting, completely relaxed, and not engaged in any activities. Each of the cases presented here indicate that the results obtained by following the proposed approach of this study are able to correctly reflect the conditions under which they were obtained.

Due to the nature of the work and time constraints involved, the scope of this project was limited to developing a functional prototype that implemented a new algorithm and demonstrated its usability. While this objective has been achieved, there are some key avenues open for future work that builds on the concepts introduced in this study. These have been identified below:

1) The approach needs to be tested on a large group of people, containing both male and female participants who cover a wider age range. This will allow the method to be tested thoroughly across a number of experiments, and will also reveal the effects of age on PWV in a more clear fashion. Moreover, the results obtained from such experiments must be compared to those produced by a commercial, clinically validated PWV monitor to verify their accuracy.

2) Given the increasing power and falling costs of mobile phones in the present day, there may be a possibility of porting the functionality of the prototype to operate completely on the phone itself as opposed to a computer, assuming that it provides satisfactory results across a range of test subjects first. This would be a big step forward and could make personalised, affordable PWV monitoring a reality. It would

require a thorough investigation and possibly several optimisations, as the procedures and calculations involved are quite processor-intensive, and though powerful, mobile phones do not yet match up to the capabilities of computers.

3) If the application is successfully ported to a mobile platform, there is also the possibility of extending its functionality to transmit the acquired information to a nominated doctor, who can then remotely investigate the user's PWV and take necessary action. Data could be sent over either the mobile network or wireless internet, comprising of the numerical results and graphs of the signals to allow for further inspection.

There may also be other opportunities for future work to be carried out based on the findings of this study, but the ones listed above are the few that emerge as a natural extension of the work presented here.

7.0   CONCLUSION

A new and unique way to estimate PWV has been presented. The methods described produced agreeable results as compared to other published results and reference values under different conditions, although further testing alongside a commercial monitor with a large group of subjects has been identified as an important step to establish its exact accuracy. However, this study has proved that it is indeed possible to process simple audio and video signals to successfully calculate PWV, which was the primary research question central to this study. Additionally, the answers to the sub-questions listed in Section 1.4 were also determined, as given below:

1) *How can the audio and video data be acquired using the mobile phone?*
   - The video data is obtained from the tip of the left index finger using the mobile phone camera and flash sensor.
   - The audio data is obtained from a portable Doppler ultrasound probe placed over the left side of the chest, connected to the mobile phone via a microphone adapter and stereo cable.
   - Data recordings are performed for a period of one minute.

2) *How can the acquired data be processed to produce information required for PWV calculation?*
   - For the audio, a 30 second sliding window is used to extract the raw audio data, which is then filtered using a band-pass filter and a Hilbert transform, the result of which is a waveform that gives the first timing reference.
   - For the video, a 30 second sliding window is used to extract and normalise the red, green and blue channels for all video frames, after which the blue channel waveform is used as the second timing reference.

3) *What algorithm can be used to compute PWV?*

- Peak and foot detection is performed on the processed audio and video waveforms, respectively.

- Ten sets of matching peaks and feet are selected from each window, and the pulse wave velocities for each set are calculated using [Eqn. 5.1]. Results outside the range of 2 to 15 m/s are considered invalid and discarded.

- The valid values are averaged to give a single average velocity for each window.

- The final average PWV is calculated at the end as the average of all velocities for each window.

Thus, by following the above processes, it is possible to compute average PWV from a given audio and video signal for a subject. While similar in principle to other methods or devices, the exact processes adopted by this study do not appear to have been investigated until now, thus presenting the possibility of a completely different and novel approach to measuring this increasingly important physiological phenomenon.

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