Blood-pressure measurement

Bimo Sunarfri Hantono

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Blood-Pressure Measurement

Bimo Sunarfri Hantono

School of Electrical & Electronic Engineering

**A thesis submitted to the Nanyang Technological University
in fulfillment of the requirement for the degree of
Master of Engineering**

2006

*To my family and friends,
who always give me love and support*

Statement of Originality

I hereby certify that the work embodied in this thesis is the result of original research and has not been submitted for a higher degree to any other University or Institution.

.....
Date

.....
Bimo Sunarfri Hantono

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Abstract

Blood pressure is one of the most common techniques used to diagnose heart disease. The blood-pressure waveform gives vital information about the blood flow through the body. In this project an auscultatory method is used to obtain non-invasive human blood-pressure signals. An algorithm based on wavelet transform is introduced to detect every single Korotkoff pulse of blood pressure. The aim is to automatically determine the systolic and diastolic points of blood-pressure measurement for (1) a subject at rest, (2) a subject walking with his/her arms still and (3) a subject walking with his/her arms swinging. With the multiscale feature of wavelet transform, pulses of blood pressure can be distinguished from noise and artifacts. The approach of the algorithm has been developed using Matlab engineering software. Detection techniques were developed to de-noise the signal, identify the blood-pressure pulses and finally to determine the systolic and diastolic points.

Chapter 1

Introduction

1.1 Motivation

Cardiovascular mortality decreased progressively in the industrialized countries in the second half of the twentieth century [37]. Improved control of hypertension has considerably contributed to this decline. For example, the Health Examination Surveys in the United States of America have demonstrated that while 10% of hypertensive persons had their blood pressure lowered below 140/90 mm Hg in the period 1976 till 1980, this proportion had risen to 27% [7] by 1988 till 1991. Still, this leaves over 70% of hypertensive persons with imperfect control, as has been reported in many countries [22] [38]. This majority of imperfectly controlled hypertensive persons reflect the inherent problem of maintaining long-term therapy for an as yet asymptomatic condition, particularly when therapy may interfere with the patient's quality of life and its immediate benefits are not obvious to all patients.

Given the aging of the population and the increased prevalence of obesity, strokes and coronary heart disease events are expected to increase or remain stable even if primary and secondary prevention of cardiovascular disease is tightened further. Given the central role of elevated blood pressure in the pathogenesis of both coronary heart disease and stroke, it is clear that one of the biggest challenges public health authorities and medical practitioners face, is the worldwide control of hypertension.

Cardiovascular risk rises linearly with blood pressure and the definition of hypertension is therefore, arbitrary. Most of the evidence about the benefits and risks of lowering blood pressure comes from studies in patients selected on the basis of high blood pressure. It is therefore not clear whether estimates of treatment effect obtained from trials in hypertensives can be extrapolated to individuals with lower blood-pressure levels [40].

As seen in Table 1, *Guidelines for the management of hypertension (World Health Organization/International Society of Hypertension 1999 and Joint National Committee VI)*, hypertension is defined as a systolic blood pressure of 140 mm Hg or greater and/or a diastolic blood pressure of 90 mm Hg or greater. Like the previous guidelines, issued in 1993, the 1999 guidelines concentrate on the management of patients with mild to moderate hypertension since there often is uncertainty among clinicians and policy makers on how to manage this condition.

Table 1. Definitions and classifications of blood-pressure levels (mm Hg)

Category	Systolic	Diastolic
Optimal	< 120	< 80
Normal	< 130	< 85
High-normal	130 - 139	85 - 89
Grade 1 hypertension (mild)	140 - 159	90 - 99
Subgroup: borderline	140 - 159	90 - 94
Grade 2 hypertension (moderate)	160 - 179	100 - 109
Grade 3 hypertension (severe)	≥ 180	≥ 110
Isolated systolic hypertension	≥ 140	< 90
Subgroup: borderline	140 - 149	< 90

Stroke

Blood-pressure levels, both systolic (SBP) and diastolic (DBP), relate positively and continuously to the risk of stroke across a wide range of levels in Western countries.

Among individuals of mostly middle age, a prolonged 5 mm Hg lower level of usual DBP was associated with a 35-40% lower risk of stroke. The slope of this association declines somewhat with increasing age [30], because the incidence of stroke increases rapidly with age. Elderly people still suffer the large majority of blood pressure related cerebrovascular disease. Blood-pressure levels are positively related to both cerebral hemorrhage and cerebral infarction, but the association appears to be somewhat steeper for hemorrhage than infarction [14]. Atrial fibrillation enhances stroke risk significantly. Stroke risk increases fourfold in all age groups once atrial fibrillation develops [42].

Sudden death

Hypertensive patients with LVH (*Left Ventricular Hypertrophy*) have a significantly greater prevalence of premature ventricular contractions and complex ventricular arrhythmias than patients without LVH and/or normotensive patients [24]. A study from the Framingham cohort [5] showed that the presence of asymptomatic ventricular arrhythmias was associated with a nearly two-fold increase in mortality.

Atrial fibrillation

Atrial fibrillation may be paroxysmal, persistent or chronic, and a number of attacks are asymptomatic. Suspicion or confirmation of atrial fibrillation necessitates investigation and, as far as possible, appropriate treatment of associated cardiovascular diseases such as hypertension, diabetes mellitus, hypoxemia,

hyperthyroidism and congestive heart failure. Hypertension is one of the major underlying conditions in atrial fibrillation [39]. In Framingham, 8% of all patients with hypertension developed atrial fibrillation in 7 years time [42].

Coronary heart disease

Blood-pressure levels have also been shown to be positively and continuously related to the risk of death due to coronary heart disease or of nonfatal myocardial infarction [21]. The increase in coronary heart disease with elevated blood pressure is about two thirds as steep as that for stroke, and appears to be similar across a broad range of blood-pressure levels, that includes both hypertensive and normotensive persons.

Heart failure and renal disease

The risk of heart failure and of renal disease relates to blood-pressure levels, but the relationships are less tight than for stroke and coronary heart disease. Nevertheless, patients with a history of hypertension have at least a six times greater risk of heart failure than individuals without such a history.

Among patients with mild hypertension, not only blood pressure is important, also the presence of other factors determine the risks of cardiovascular disease. For example, a man aged 65 years with diabetes, a history of transient chemical attacks and a blood pressure of 145/90 will have an annual risk of a major cardiovascular event that is more than 20 times greater than in a man aged 40 years with the same blood pressure but without either diabetes or a history of cardiovascular disease. In contrast, a man aged 40 years with a blood pressure of 170/105 mm Hg will have a risk of a major cardiovascular event that is about two or three times greater than that of a man of the same age with a blood pressure of

145/90 mm Hg and similar other risk factor levels. Therefore, differences in the absolute level of cardiovascular risk between patients with hypertension is determined to a greater extent by other risk factors than by the level of blood pressure.

The above illustration shows that the blood-pressure measurement is one of many important physiological measurement used to determine the health condition of human being.

1.2 Objectives

The objectives of our project were to automatically determine the systolic and diastolic pressure measurements for a subject at rest, a subject walking with his/her arms still and a subject walking with his/her arms swinging. Wavelet transform was employed to process the blood-pressure signals obtained using the auscultatory method. Detection techniques were developed to de-noise the signal, identify the blood-pressure pulses and to eventually determine the systolic and diastolic points.

Most automatic blood-pressure measurement requires the subjects to be stationary. Thus the results from our project can be applied to devices, such as the Ambulatory Blood Pressure Monitor (ABPM) [46] [47], an event-driven, patient-activated blood-pressure monitor, and also can be used on a person who is exercising on a treadmill during stress test or in a gymnasium. Here the subjects are expected to be non-stationary and moving.

1.3 Major Contribution of the Thesis

A literature review was first conducted and many ideas were generated after studying the references and journal papers. Many methods have been proposed based on Wavelet transform in the past decades. Biomedical-signal analysis is one of the most frequent use of Wavelet transform, since this method is very promising in detecting the transient-signals that are usually found in biomedical signals.

During extensive study for the project, the following contributions were made. First, the various current methods to determine the blood pressure were studied. Methods have been observed and analyzed to figure out which one is better. Second, the performance of different Wavelet transform for pulse detection was also studied. The principle of the quadratic-spline Wavelet transform that have been done in [10] and [11] were reviewed. An algorithm was then proposed to detect the Korotkoff blood-pressure pulses that were obtained from the auscultatory measurement. The disadvantage of the quadratic-spline Wavelet transform is its heavy computational cost. In order to reduce the problem, study of the Korotkoff blood-pressure pulse itself was done, to understand the characteristics of the blood-pressure pulse. This can help to reduce the computational cost and a good performance can be obtained. Wavelet analysis is a new signal analysis method.

Unlike Fourier analysis, by which we can know the message in frequency domain, wavelet analysis can let us know both time domain and frequency domain messages. Thus, it is very useful for analyzing transient signals.

1.4 Organization of the Thesis

The thesis has been organized into 5 chapters follows:

Chapter 1 Introduction

This chapter introduces the background and the outline of the project, the objectives it has to fulfill and the scope by which it was bounded.

Chapter 2 Blood-Pressure Measurement

This chapter gives the basic concept of blood-pressure measurement and describes various types of blood-pressure measurement techniques currently in use.

Chapter 3 Theory

This chapter discusses the blood-pressure measurement methodologies being used in the project. Some previous project in this field will also be briefly discussed. This chapter also discusses the wavelet theory and how the wavelet provides an elegant methodology to differentiate between the “real” waveform and noise in a biomedical signal. The use of wavelet transform in signal processing is discussed here.

Chapter 4 Implementation

This chapter describes the implementation of the blood-pressure measurement algorithm that has been proposed in this project.

Chapter 5 Conclusions and Recommendations

This chapter concludes the entire project and proposes some further enhancements to the project.

Chapter 2

Blood-Pressure Measurement

Blood pressure is probably the next most frequent measured biomedical quantity beside the body temperature. Blood pressure is the most common index of cardiovascular performance utilized today [43]. It is used in the diagnosis of a wide variety of complications that are associated with the cardiovascular system. The development of different methods and characterizations of the blood-pressure waveform are very important in the ability to more quickly and accurately characterize these complications.

2.1 Systole and Diastole

The heart's pumping cycle is divided into two major parts: the *systole* and the *diastole points* [20]. The systole is defined as the period of contraction of the heart muscles, specifically the ventricular muscle, at which time blood is pumped into the pulmonary artery and the aorta. The diastole is the period of dilation of the heart cavities as they fill with blood. Once the blood has been pumped into the arterial system, the heart relaxes, the pressure in the chamber decreases, the outlet valves close, and in a short time, the inlet valves open again to restart the diastole and initiate a new cycle in the heart.

The heart beats at an average rate of 75 beats per minute in a normal adult. The heart rate increases when a person stands up and decreases when he sits down,

the range being 60 to 80. On the average, it is higher in women and generally decreases with age. The systolic (maximum) blood pressure in the adult is in the range from 95 to 140 mm Hg, and the average is 120 mm Hg. The diastolic blood pressure ranges from 60 to 90 mm Hg, and the average is 80 mm Hg. These figures are subject to variation with age, climate and eating habits.

2.2 Blood-Pressure Monitoring Methods

Blood pressure may be measured using direct (*invasive*) or indirect (*non-invasive*) techniques. Direct measurements use catheters to invasively determine blood pressure, whereas indirect methods utilize a variety of non-invasive techniques.

Direct measurements utilize one of two different types of sensors, in conjunction with a catheter, to determine blood pressure. Extra vascular sensors are located outside of the body and use the principle of wave propagation to transmit vascular pressure from the measurement site to the sensor via a fluid-filled catheter. In contrast, intra vascular sensors, such as fiber-optic pressure sensors, are positioned on the tip of the catheter and inserted into the artery. Catheter systems are often used for direct, continuous measurement of intra-arterial blood pressure in the aorta, which is considered the “gold standard” for blood-pressure measurement.

The need for this type of continuous, highly accurate reading of a patient’s central blood pressure is commonly found in critical care surgical settings. However, the invasive nature of catheter procedures prohibits them from being a viable option for routine blood-pressure measurement. Instead, physicians use non-invasive

methods for casual measurements. Therefore, a major goal in the field of non-invasive blood-pressure monitoring has been the development of a method, which correlates with “gold standard” catheter measurements. The non-invasive measurement of blood pressure may be performed in several different ways. Many methods use an air-filled cuff to temporarily occlude blood flow through the artery, and then apply a particular technique to obtain blood-pressure data while the cuff deflates. The most common indirect technique is the auscultatory method, in which a clinician determines systolic and diastolic pressures by listening to the characteristic Korotkoff sounds of the blood flow in the transducer artery during cuff deflation. The Korotkoff sounds have been categorized into five phases (descriptions taken from ANSI/AAMI 1987 guidelines). See Table 2.

Table 2. Five phases Korotkoff sound

Phase I	Begins with the sudden appearance of a faint, clear, tapping or thumping sound that gradually increases in intensity.
Phase II	Phase I ends and phase II begins when the sounds change to a loud “swishing” murmur.
Phase III	The beginning of Phase III occurs when the sounds assume a loud, distinct, knocking quality. These sounds are less intense than those of Phase I.
Phase IV	Begins when the sounds suddenly become muffled and have a faint murmur-like or “swishing” quality.
Phase V	Begins when silence develops.

The beginning of Phase I corresponds to Systolic Blood Pressure (SBP). The actual point of Diastolic Blood Pressure (DBP) based on these Korotkoff sounds is less clear. The ANSI/AAMI 1987 report suggests that the pressures at the beginnings of both phases IV and V be recorded, since there may be some debate as to which value best represents DBP. The phase IV values tend to be higher than catheter measurements, while phase V values are usually lower. The DBP criteria is further

complicated by the fact that some patients may not have audible phase IV sounds, whereas in others, the beginning of phase V (silence) may be difficult to determine. Since the auscultatory technique is based on the ability of the human ear to detect and distinguish sounds, there is a possibility for measurement error due to individual levels of auditory acuity and sensitivity. Although a fully-qualified clinician can consistently obtain accurate blood-pressure measurements, unqualified or inexperienced personnel may be more susceptible to outside noise interference, or inconsistent assessment of the actual points of Phase I and Phase IV or V. In an attempt to increase reproducibility, some automated devices have replaced the human ear with a microphone. These devices apply sound-based algorithms to estimate SBP and DBP.

In addition to noise-artifact sensitivity, these sound-dependent algorithms may not adequately compensate for patient conditions such as hypotension (i.e. low blood pressure), where the Korotkoff sounds may be muted. There is no way to determine mean arterial pressure (MAP) solely by the use of Korotkoff sounds, and that is a strong limitation of auscultation. In order to provide an estimation of MAP a formula has been developed, which is quite commonly used in auscultatory devices [43]:

$$MAP = \frac{SBP}{3} + \frac{2 \cdot DBP}{3} \quad (2.1)$$

Oscillometric blood-pressure determination is another common non-invasive method of blood-pressure monitoring. The term “oscillometric” refers to any measurement of the oscillations caused by the arterial pulse. These oscillations are the direct result of the coupling of the occlusive cuff to the artery. This technology

was originally designed as an alternative to the auscultatory technique, allowing blood-pressure measurement of critical care and intensive care unit (ICU) patients whose Korotkoff sounds were inaudible (usually due to hypotension caused by massive hemorrhage or shock).

In oscillometric techniques, the cuff is first inflated until the artery is fully occluded. Then, the monitor takes measurements while the cuff deflates. Most oscillometric devices examine the pulsatile pressure generated by the arterial wall as it expands and contracts against the cuff with each cardiovascular cycle. The pressure transducer based on the distension of the artery generates an electrical signal. Over the course of the measurement process, the magnitude of the pulsatile signal increases, reaches maximum amplitude, and then decreases.

A common algorithm used by many oscillometric devices sets MAP equal to the point of maximum amplitude. SBP and DBP are then determined by the application of predetermined systolic and diastolic ratios. In the height-based approach, for example, ratios of pulse amplitude to maximum amplitude are used. For this type of algorithm, the maximum amplitude is defined as MAP.

An oscillation that satisfies the systolic ratio and occurs before the MAP point is considered SBP. An oscillation that satisfies the diastolic ratio and occurs after the point of MAP determination is considered DBP. The determination of both systolic and diastolic ratios are based on the correlation to auscultatory or catheter measurements. Unfortunately, an erroneous determination of MAP, due to any form of artifact, may produce inaccurate values for SBP and DBP.

The majority of the monitors on the market today are either auscultatory or oscillometric in nature [45] [28] [19], however, there are other types of devices. In addition, some monitors employ both auscultatory and oscillometric methods, using one method as a primary measurement and the other as a secondary measurement for verification.

Not all techniques yield both SBP and DBP. For example, the palpatory method only measures SBP [45] [28] [19]. In this technique, the artery is occluded with a cuff. Then, the cuff is allowed to deflate, and SBP is determined by measuring the cuff pressure at which a radial or finger pulse is first detected. Most techniques, however, provide both SBP and DBP. The infrasound method, for example, attempts to improve on the auscultatory method by detecting low frequency Korotkoff vibrations below about 50 Hz, i.e. including sub-audible vibrations.

The ultrasound method is quite different from infra sound, because it measures Doppler shifts to determine blood pressure. Ultrasound waves generated from a transmitter located distal to the cuff are projected towards the artery. These waves contact the arterial wall and are reflected back to the receiver. Distension of the wall causes phase shifts in the reflected waves (known as Doppler shifts). From these phase shifts the opening and closing of the artery can be determined. Systolic pressure is the point at which the artery can remain open with the largest cuff pressure. A similar algorithm determines diastolic pressure.

Another method, known as impedance plethysmography [45], also measures the volumetric change associated with arterial distension. The volume changes cause changes in the electrical conductivity (impedance) of the measurement site. The pulsatile nature of the volume changes (due to the cardiovascular cycle) is

reflected in the measurement of impedance pulses. When graphed over time, these pulses produce a waveform similar to the pressure-generated oscillometric waveform. Pressures are then estimated in a manner similar to the oscillometric technique.

Arterial tonometry utilizes a very different approach [45] [28]. The artery is flattened and the pressures required to maintain that flat region are measured. This is accomplished by using an array of sensors, each of which measures pressure. This array ensures that the maximal pressure being exerted on the arterial wall by the blood is felt. The result of this method is a waveform similar to catheter measurements, and some type of algorithm must be used to calculate pressures from that waveform.

There are several limitations to tonometry. First, it is a measure of the peripheral circulation, with much different value for pressure that is felt closer to the heart. A second limitation of tonometry is its high sensitivity to sensor location and angle. With each relocation of the sensor, there can be varying values for pressure. Experienced, well-trained operators may minimize measurement errors. However, inter-operator reproducibility still lacks consistency. Lastly, tonometry requires an independent blood-pressure measurement technique for calibration.

2.3 Comparison of Blood-Pressure Measurement Methods

Invasive methods of blood-pressure measurement are very accurate, but are rarely used. Non-invasive methods are far more popular due to their ease, relative safety to the patient and cost. Some of the non-invasive methods that are currently

utilized are the auscultatory measurement and the oscillometric measurement methods [31]. The most popular of these methods is the auscultatory method due to its ease of application, speed, and cost.

In the comparison of the methods mentioned above, we were lead to investigate and use the well-proven and accepted auscultatory method in this project. The clinically-proven auscultatory technique of blood-pressure measurement is widely used by clinicians everyday with good accuracy in all patient groups. Some of the considerations made were as follows.

1. **Cost.** Amongst the techniques, the auscultatory method is the cheapest, and easiest to set up. Next come the oscillometric method, and the ultrasound method.
2. **Ease of use.** The auscultatory method is also the easiest to use. Because there is no sophisticated equipment involved, and a hassle-free set up, there was absolutely no problem in implementing the auscultatory method. Oscillometric and ultrasound methods require more precise set ups. Moreover, recording of signal was simple for the auscultatory method. In our project, the signal obtained can be simply captured by the "line-in" of the sound card of a personal computer and recorded as a wave file.
3. **Effectiveness.** Both the auscultatory and the ultrasound methods are able to provide the systolic and diastolic pressure points. The oscillometric method, however, can only attain the mean arterial pressure.

4. **Acceptability.** The auscultatory method is the most commonly accepted method. It is widely used in the clinics, hospitals and treadmill stress tests. Most importantly the American Heart Association (AHA) endorses it.

2.4 Automatic Blood-Pressure Measurement

Since blood pressure is generated by cardiovascular activity, it is accepted worldwide as an index of cardiovascular performance. Therefore, impairment of cardiovascular function may often be detected through blood pressure changes. However, ease of determination is an additional factor, which contributes to the acceptance of blood pressure as an index of cardiovascular performance.

Any type of measurement that is to be performed on human subjects is governed by its comfort, convenience and necessity. A procedure that is deemed impractical for routine use is of limited value to a physician. Invasive catheter procedures can obtain accurate blood-pressure measurements in both central and peripheral locations of the body, providing the greatest flexibility with regard to measurement location. However, such measurements require surgical insertion procedures, and are seldom utilized for routine blood-pressure measurements. Instead, non-invasive techniques are usually utilized for most cases. These non-invasive measurement techniques usually utilize a peripheral measurement site, and do not supply measurement information that directly pertains to the central circulation.

Due to current advances in technology, blood pressure may be measured using quick, convenient, non-invasive procedures that facilitate its integration into routine examinations. Because of the limitation of manual blood-pressure measurement, which usually necessitates a trained personnel, at this time there are many type of automatic blood-pressure measurement commercially available in the market with various methods and algorithms.

A number of automatic blood-pressure measuring systems of sphygmomanometer type have been developed. The basic technique includes an automatic sphygmomanometer that inflates and deflates a pneumatic cuff at a predetermined rate (deflate at rate of 2 to 3 mm Hg/sec). A sensitive detector is used to measure the distal pulse. Different kinds of detectors have been employed including ultrasonic, photoelectric, oscillometric or piezoelectric. Pulse transducers of the piezoelectric type are commonly used.

Most automatic blood-pressure measurement systems presently found in the market utilize the oscillometric measurement principle. A transducer detects the orderly sequence of oscillations in cuff pressure as it is reduced from above systolic to below diastolic pressure. With a supra systolic cuff pressure, the pressure indicator show small oscillations. When cuff pressure fall below systolic, the oscillations increase in amplitude. With a continued reduction in cuff pressure, the oscillations increase further; sometimes they remain constant in amplitude. However more often they continue to increase, then decrease. In this spectrum, there is an identifiable maximum point for oscillations which is approximately the MAP.

However, most of the automatic blood-pressure measuring systems only work well within an acceptable margin on healthy stationary subject, sitting down or lying down, but fail when used to measure blood pressure during activity, for example, when walking on a treadmill during a stress test. Even the slightest movement would cause the devices to reflect an erroneous result or totally fail to give a reading.

2.5 Problem Associated with Blood Pressure Measurement of Non-Stationary Subject

Normally, the measurement of blood pressure using the auscultatory method is done on the patient either sitting on the chair or lying on the bed. With this method, the doctor can easily determine the systolic and diastolic points of blood pressure from a stationary subject.

Figure 1 shows samples of Korotkoff blood-pressure waveform from stationary and non-stationary subject respectively. The blood pressure was measured using the auscultatory method. The blood-pressure pulses were picked up by a piezoelectric transducer placed at the brachial artery to listen to the Korotkoff sounds. The waveforms were plotted using Matlab software.

By observing the magnitude of the signal pulses from Figure 1(a), the blood-pressure pulses from stationary subject, the doctor can determine which is the systolic and the diastolic points either from the figure or direct from the stethoscope. In these cases the trained personnel can hear the recorded Korotkoff sound. The systolic value (S) occurred when the first sharp dominant pulse

appeared and the diastolic value (D) occurred when the last sharp dominant pulse disappeared.

The second figure, Figure 1(b), shows a sample of noisy blood-pressure waveform taken from a subject walking or running slowly on the treadmill. As it can be seen, the blood-pressure pulses cannot be differentiated from noise spikes caused by subject motion. Determination of the systolic and diastolic points are thus not possible by simple signal amplitude discrimination. Moreover, the noise caused by the walking motion muffles the blood-pressure pulses. It would be considerable more difficult, if not impossible, for a medical doctor using a stethoscope to measure the blood pressure of subjects in motion.

Figure 1. Blood-pressure waveform samples: (a) stationary subject, and (b) non-stationary subject

Thus, automatic detection of systolic and diastolic values are complicated by the introduction of noise, as a result of moving artifact while the person is exercising on the treadmill. These noise spikes co-exist with the actual blood-pressure pulse

spikes. The difficulty is made more complicated by the fact that these noise spikes are very random in occurrence and it is very difficult to predict when and where they will occur. It is impossible to design a low-pass filter with an exact cut-off frequency to filter away all the unwanted noise signals.

The objectives of our project were to determine systolic and diastolic pressure measurements for a subject at rest, a subject walking with his/her arms still and a subject walking with his/her arms swinging. Wavelet transform was employed to process the blood-pressure signals obtained using the auscultatory method. Detection techniques were developed to de-noise the signal, identify the blood-pressure pulses and then to eventually determine the systolic and diastolic points.

Chapter 3

Theory

3.1 Previous Work on Blood-Pressure Measurement

There were very few literature or research on blood-pressure measurement unlike other biomedical areas, such as electrocardiogram (ECG) and diagnostic imaging. There are some literature associated with blood-pressure measurement: [4] and [18] deal with the oscillometric method; [31], [23] and [26] used ultrasound detector method; [8] used statistical and fuzzy model with ECG data approach to determine systolic point; [41], [27], [32] and [33] were working on palpation method. However, signal processing technique used in other biomedical areas are insightful and they can be adapted for our blood-pressure measurement.

Wavelet transform has been successfully applied to physiological signal such on ECG to detect their characteristic points [10]. The wavelet transform can also be applied to each blood-pressure pulse as they have characteristics similar to ECG.

3.2 The Wavelet Transform

The wavelet transform method is similar to the well known Fourier Transform and Short Time Fourier Transform (STFT) method [2] [1], although it is not as easily described in mathematical terms as the Fourier Transform method since the basis function, or mother wavelet, is far more complicated than the basis functions,

sines and cosines, used in the Fourier method. Furthermore where the STFT provides signal decomposition into a set of equal bandwidth functions in the frequency spectrum, the wavelet transform provides decomposition of a signal into a set of constant Q band-pass functions, i.e. providing equal bandwidth on a logarithmic scale.

Other conditions that the mother wavelet may comply with, such as orthonormality and compact support, may also be desired [2]. The mother wavelet is used to form a set of wavelets by scaling (dilating or compressing) and translating (shifting) the mother wavelet.

A set of wavelets can be obtained from the general expression $\frac{1}{s}\psi\left(\frac{x-t}{s}\right)$,

where s is the scaling factor and t the translation term and $\psi(x)$ the mother wavelet function. The continuous wavelet transform of a signal function $f(x)$, with respect to a mother wavelet, $\psi(x)$, can be defined as:

$$W_s f(x) = f(x) * \psi_s(x) = \frac{1}{s} \int_{-\infty}^{+\infty} f(t) \psi\left(\frac{x-t}{s}\right) dt \quad (3.1)$$

A wavelet coefficient, W_s at a particular scale and translation represents how well the signal, $f(x)$, and the scaled and translated mother wavelet match or correlate. The set of all wavelet coefficients, $W_s f(x)$, is the wavelet domain representation of the function $f(x)$ with respect to the mother wavelet $\psi(x)$.

The wavelet transform can be related to the Fourier transform [1]. Wavelet

methods represent functions as a weighted sum of scaled and translated mother wavelets. The mother wavelet replaces the exponential term in the Fourier method and scale and translation replace frequency shifting. Finally, a two-dimensional surface of wavelet coefficients replaces the one-dimensional Fourier coefficients.

With a mother wavelet chosen to be $\psi(x) = e^{jx}$, $s = 1/\omega$, and $t = 0$, then the wavelet transform in equation (3.1) becomes:

$$W_{\psi} f(s, t) = W_{e^{jx}} f\left(\frac{1}{\psi}, 0\right) = \int f(x) e^{-j\omega x} dx = F(\omega) \quad (3.2)$$

which is the Fourier transform [1] [11] [29].

The Wavelet Transform is linear, i.e. if forming a weighted sum of separate signals creates a composite signal then the wavelet transform of the composite signal is equal to the weighted sum of the wavelet transforms of each separate signal.

3.2.1 Discrete-Time Wavelet Transform (DWT)

The independent variable of the function to be approximated by wavelet decomposition is discretized. The discrete-time wavelet transform maps discrete finite energy sequences to a 2-D discrete grid of coefficients.

A specific structure can be implemented, as in references [1] [15], where time dilation is accomplished by simply dropping every other sample of the signal under analysis for every stage in the wavelet decomposition, see Figure 2. The signal is convolved with the impulse response of the low-pass and high-pass filter. The decimation blocks relate to the resolution to be obtained for a given bandwidth, to produce an octave-band representation of the signal.

Figure 2. Implementation of the wavelet transform by high-pass and low-pass filtering stages with decimation

The scale parameter controls the rate of decay of the mother wavelet, i.e. controls the size of the analysis window dictated by the mother wavelet. The Fourier method, on the other hand, changes the frequency of the oscillating exponential under a fixed duration window, but does not change the duration of the window. This results in a fixed-resolution in the time-frequency plane, which does not change as the frequency changes. The wavelet transform is especially suitable for blood-pressure analysis applications, which require wide-band or non-stationary signal processing.

The close relationship between perfectly reconstructing filter banks and dyadic wavelets lead to computational efficiency [13] [17] [25]. In this method the wavelet transform is implemented with a series of cascaded low-pass filters and high-pass filters whose corresponding frequency responses are separated by octaves.

The proposed method for blood-pressure measurement analysis would initially make use of the wavelet transform for the detection of each Korotkoff pulse in the blood-pressure measurement.

The proposed dyadic wavelet transform used in [10] and [2], is a *quadratic-spline wavelet* with compact support. This can also be used for extracting blood-pressure characteristic points. The filters used to implement the Mallat algorithm [35] provide a wavelet that has compact support [3], or, in the other words the duration of the inter-scale coefficients is finite. The dyadic wavelet transform of a digital signal $f(n)$ can be calculated with Mallat algorithm [35] as follows:

$$\begin{aligned} S_{2^j} f(n) &= \sum_{k \in Z} h_k S_{2^{j-1}} f(n - 2^{j-1} k) \\ W_{2^j} f(n) &= \sum_{k \in Z} g_k S_{2^{j-1}} f(n - 2^{j-1} k) \end{aligned} \quad (3.3)$$

where S_{2^j} a smoothing operator and $S_{2^0} f(n) = d_n$. The variable d_n is the digital signal to be analysed [34], which is the blood-pressure signal used in this project. The variable $W_{2^j} f(n)$ is the Wavelet Transform of the digital signal $f(n)$. Both of $\{h_k | k \in Z\}$ and $\{g_k | k \in Z\}$ are coefficients of a low-pass filter $H(\omega)$ and a high-pass filter $G(\omega)$, respectively; that means

$$\begin{aligned} H(\omega) &= \sum_{k \in Z} h_k e^{-ik\omega} \\ G(\omega) &= \sum_{k \in Z} g_k e^{-ik\omega} \end{aligned} \quad (3.4)$$

The Figure 3 shows the comparison example from four different wavelet transform. The sample pulse and its wavelet transform were plotted by using Matlab software. The Haar, Daubechies and Symlet wavelet transform are generated by wavelet function available on the Matlab software.

Figure 3. The comparison between four different wavelet transforms

The Haar wavelet transform have similar result as the quadratic-spline wavelet transform, but the amplitude value of the Haar wavelet transform is smaller than that from the quadratic-spline wavelet transform and the differentiation of the amplitude values between each subsequent scale is not as much.

The wavelet we used is the quadratic-spline wavelet with compact support and one vanishing moment. It is a first derivative of a smooth function [34]. The Fourier transform of $\psi(x)$ is

$$\hat{\psi}(\omega) = i\omega \left(\frac{\sin\left(\frac{\omega}{4}\right)}{\frac{\omega}{4}} \right)^4 \quad (3.5)$$

The symbol $\wedge$ represents the discrete Fourier transform. For a quadratic-spline wavelet with compact support, the band-splitting filters are $H(\omega)$ and $G(\omega)$ which can be expressed as

$$\begin{aligned} H(\omega) &= e^{i\omega/2} \left(\cos \frac{\omega}{2} \right)^3 \\ G(\omega) &= 4 e^{i\omega/2} \left(\sin \frac{\omega}{2} \right) \end{aligned} \quad (3.6)$$

for the low-pass filter and high-pass filter respectively. The wavelet transform is computed with Mallat algorithm as in [10].

3.2.2 Equivalent Filter of Wavelet Transform

The discrete Fourier transform of the Wavelet Transform is

$$\begin{aligned} \hat{W}_{2^j} f(\omega) &= \hat{f}(\omega) \hat{\psi}(2^j \omega) \\ \hat{W}_{2^j} f(\omega) &= \begin{cases} G(\omega) \hat{f}(\omega) \hat{\phi}(\omega), & j=1 \\ G(2\omega) H(\omega) \hat{f}(\omega) \hat{\phi}(\omega), & j=2 \\ G(2^{j-1}\omega) H(2^{j-2}\omega) \dots H(\omega) \hat{f}(\omega) \hat{\phi}(\omega), & j>2 \end{cases} \end{aligned} \quad (3.7)$$

where ϕ is a smooth function, and $\hat{f}(\omega) \hat{\phi}(\omega)$ is the discrete Fourier transform of the digital signal $d_n (n \in \mathbb{Z})$ [34].

From the above equation, the wavelet transform of the digital signal, $f(n)$, at scale 2^j is equal to a filtered signal of d_n that passed through a digital band-pass filter (or high-pass filter for scale 2^1). Let $Q^j(\omega)$ be the transform function of the equivalent filter. Then, we have

$$Q^j(\omega) = \begin{cases} G(\omega), & j=1 \\ G(2\omega)H(\omega), & j=2 \\ G(2^{j-1}\omega)H(2^{j-2}\omega)\dots H(\omega), & j>2 \end{cases} \quad (3.8)$$

From the above equations (3.6) to (3.8), we can deduce

$$Q^j(\omega) = \frac{2^{2^j+2^{j-1}-2}}{8^{j-1}} \sum_{k=1-2^{j-1}}^{2^j+2^{j-1}-2} q_k^j e^{ik\omega} \quad (3.9)$$

$$q_{1-2^{j-1}+k}^j = -q_{2^j+2^{j-1}-2-k}^j \neq 0,$$

$$k \in [1-2^{j-1}, 2^j+2^{j-1}-2]$$

where $Q^j(\omega)$ is a finite impulse response (FIR) digital filter with generalized linear phase [12]. The filter is antisymmetric and the time delay of its central point is

$\frac{2^j-1}{2}$ (the delay is approximated as $2^{j-1}-1$ points). According to the form of the

quadratic-spline wavelet [35], every uniphase wave corresponds to a positive-negative-minimum pair of $W_{2^j}f(n)$ at different scales. The wave rising edge corresponds to a negative minimum, and the dropping edge corresponds to a positive maximum at different scales. The modulus of these maxima or minima corresponding to the same edge is named as the modulus maximum line. If the uniphase wave is symmetric to its peak, as shown in Figure 4 (a), then its peak

corresponds to the zero-crossing point of the positive-maximum negative-minimum pair with a delay of exactly $2^{j-1} - 1$ points. If the uniphase wave is not symmetric to its peak, as shown in Figure 4(b), then the peak corresponds to the zero-crossing point of the positive-maximum negative-minimum pair with a delay of about $2^{j-1} - 1$ points. The larger the scale, the bigger is the error between the delay and $2^{j-1} - 1$. At a small scale 2^1 , the error is zero. At large scale the error is still tolerant. For example, in Figure 4(b), the slope of rising edge is three times larger than that of the dropping edge, but the error is only two points at scale 2^4 . From equation (3.9), we can see that the beginning point of q_k^j is at $1 - 2^{j-1}$, and the ending point of q_k^j is at $2^j + 2^{j-1} - 2$. So, the onset of a uniphase wave corresponds to the onset of a positive-maximum negative-minimum pair with a delay of $1 - 2^{j-1}$ points, and the offset corresponds to an offset with a delay of $2^j + 2^{j-1} - 2$ points, as shown in Figure 4(e).

Figure 4. The relation between the characteristic points of simulation waves (signal) and those of their Wavelet transforms at different scales (scale 2 to scale 5)

Chapter 4

Implementation

In this chapter we will introduce each step of the algorithm that has been implemented in the project. Detection techniques were developed to identify the pulses and then to determine the systolic and diastolic points.

4.1 Detection Algorithm

Automatically measuring blood pressure accurately and non-invasively is not an easy engineering task in the best circumstances. Physiological differences between individuals and difficulty in automatically characterizing Korotkoff sounds create problems in developing a machine that performs well on all subjects. Motion or acoustic artifact during blood-pressure measurement greatly magnifies the problem [9].

Automatically measuring blood pressure can be done with various blood-pressure detection methods, as we have described in previous chapters. From all of those advantages and disadvantages, we choose auscultatory method in our project due to its ease of application, speed, cost and the most importantly it is endorsed by the medical community.

In this project, a new framework is proposed using the wavelet transform approach. The wavelet transform is introduced, as a novel method to analyze

signal in both time and frequency domains, and therefore it is suitable for analysis of time-varying signals such as blood pressure. The wavelet transform method has been successfully used for ECG characterization [10] and [11]. We used a similar method to detect the blood-pressure pulses. By using quadratic-spline wavelet transforms that have a biphasic feature, every uniphase wave corresponds to a positive-maximum negative-minimum pair of wavelet coefficients. We can then easily determine each single Korotkoff pulse of blood pressure.

The algorithm designed to detect systolic and diastolic pressure points for a subject at rest, a subject walking with his/her arms still and a subject walking with his/her arms swinging is divided into three parts:

1. **Peak detection.** This is the first step in the algorithm to determine the starting point and to detect the first possible blood-pressure pulse. Peak detection of the envelope of the Korotkoff pulse train is used to find the starting point of the detection process. The highest-amplitude Korotkoff pulse occurs in the middle of the Korotkoff pulse train. Here the signal-to-noise ratio is the highest. We will first detect the peak pulse and proceed to detect the rest of the Korotkoff pulses from the middle point toward the starting and end point of the pulse train.
2. **Pulse detection.** In this step, the quadratic-spline wavelet transform will be used in the main algorithm to determine each Korotkoff blood-pressure pulse. This is the most important step in the algorithm. The algorithm for this step has been designed to increase the probability of detection.

3. **Systolic and diastolic points determination.** This step is to determine the systolic and diastolic pressure points from the blood-pressure pulse train detected from the previous step.

The next few sections will describe in more detail the proposed algorithm that has been developed. Part 1, 2, and 3 are described in the following Sections 4.3, 4.4 and 4.5 respectively. Section 4.2 focuses on the selection of the threshold values that are used in the pulse-detection algorithm described in Section 4.4.

4.2 Selection of Threshold Value

This section will discuss how to determine the threshold value that will be used in the determination of each blood-pressure pulse which is described in Section 4.4. The threshold value is needed to determine whether the analyzed waveform is a pulse or not. Three standard blood-pressure pulse train characteristic curves [6] are shown in Figure 5.

The blood-pressure pulse varies in amplitude at different positions, as shown in Figure 5(a) and (b). That is why a constant threshold is not recommended in the blood-pressure pulse detection process. Hence adaptive threshold will be used in the detection algorithm. Figure 5(c) rarely occurs and usually happen on subjects in the uncommon condition, e.g. under treatment or hypertens. The first two waveforms are considerably more common.

Since the threshold should be made adaptive to include all possible cases of blood-pressure pulses, the threshold value will be updated after we successfully detect a blood-pressure pulse. Now we start our detection process from the center

of the pulse train, moving toward the front and tail of the pulse train. From our observations of blood-pressure waveform samples, it was seen that the average decreasing amplitude of the next/previous pulse is 78% of the amplitude of the current detected pulse, with standard deviation of 25.5%. Hence, the new threshold will be set to 78% of the amplitude of the current detected pulse. This threshold value will be used for the next-pulse detection.

Figure 5. Three most common blood-pressure pulse train curves and the possible location of the systolic and diastolic point

In our proposed algorithm, if the blood-pressure pulse is not detected within the detection range with the current threshold value, the threshold will be further reduced another 78% and the detection process will restart again. When the arms are swinging the variation of amplitude between pulses may vary as much as 40%, though the main envelope is still maintained. That is where this procedure comes into play. Under most situations, changing to 78% of the amplitude of current pulse is enough. This procedure is given the first priority in detection.

4.3 Peak Detection

Under normal circumstances, the blood-pressure pulse will vary in amplitude as the pressure changes as shown in Figure 5(a) and (b). The noise introduced into the system will not vary much with pressure. Hence, the highest SNR occurs when the amplitude of blood-pressure pulses is highest. In any system, it is always easier to separate signal from noise with high SNR (signal-to-noise ratio) than low SNR. Hence, in this algorithm, the peak of the envelope is chosen as a starting point for the detection process. We then proceed to detect the pulses starting from this central point working toward the front of the pulse train and then toward the rear of the pulse train.

The main function of this part, therefore, is to obtain this central point and save the corresponding sampling point as the starting point for further detection. The algorithm to obtain this point is shown in Figure 6. From this flowchart, we see that initially the waveform is passed through a band-pass filter with bandwidth 10 Hz -180 Hz. The bandwidth 10 Hz -180 Hz is chosen mainly because almost all energy of the blood-pressure signal will fall within this range (as described in Section 4.4.3). Hence, noise energy will be reduced and overall SNR will improve. Besides that, the main function of the band-pass filter in this case is to attenuate impulse noise that has high amplitude. This is necessary because the high-amplitude impulse noise that is present may appear as the highest peak, and wrong peak detection will occur.

Figure 6. Flowchart of obtaining envelope peak

After passing through the band-pass filter, the peak of the envelope is detected by using the **max()** function. This function was done within a central window with a size of 60% of the total blood-pressure waveform. The first 20% and last 20% of the pulse train are not considered. This central window size is necessary because some rare waveforms have almost constant envelope as shown in Figure 5(c). If no central window size is used, the maximum point may be the first blood-pressure pulse at the head of the pulse train (if it happens to be the highest amplitude), which is undesirable. This is why 20% of the starting and trailing sample points were excluded, so as to prevent this kind of inaccurate detection. Although this kind of waveform is rare, the probability of it happening exists; hence this safety precaution must not be excluded.

4.4 Korotkoff Blood-pressure pulse detection

4.4.1 Pulse-Detection Approach

Blood-pressure pulses can be verified using the wavelet detection techniques. It was observed during the project, that the shape of blood-pressure pulses varies from person to person. This variance in shape can even take place from the systolic pressure point down to the diastolic pressure point of the same person. From the several blood-pressure signal waveforms that had been obtained, five prevalent blood-pressure pulses and its quadratic-spline wavelet transform signal for scales varying from $j = 2$ to $j = 5$ were selected and displayed as shown in Figure 7.

Figure 7. Five prevalent Korotkoff blood-pressure pulses and their Wavelet Transform at different scale (scale 2 to scale 5)

4.4.2 Verification of Wavelet Transform Techniques

Some of the characteristics of the quadratic-spline wavelet transform technique for detection of blood-pressure pulses are as follows (as shown in Figure 4) :

1. The increasing scale of the wavelet transform would increase the amplitude of the blood-pressure wavelet transform coefficients. This correlation would stay true up to a certain scale level (mostly scale 5), and beyond that, the amplitude of the wavelet transform coefficients would start to decrease. For example in Figure 4, the modulus of the maxima or minima increases as the scales increase from $j = 2$ to $j = 5$.
2. In the various scales of wavelet transform, the wavelet transform coefficients are biphasic, with a negative peak following the maximum peak or the maximum peak following the minimum for all the different scales, but it is consistent for all scales.

From Figure 7, one important observation can be made. Although the shape of blood-pressure pulse varies, its wavelet transform characteristic remains largely the same. The wavelet function is in effect a band-pass filter. Most of the energy of the blood-pressure pulse are at scales of $j = 3$ and $j = 4$. Typically the coefficients increase and then decrease across 3 or 4 consecutive scales where most energy are present. In addition, it is observed that most of the blood-pressure pulses have wavelet-transform coefficients with their minimum peaks occurring before maximum peaks. Using this observation, the algorithm is set to detect for a minimum peak first, followed by the maximum peak. However if no pulses are found, it will then be set to detect for maximum first then the minimum. This helps to speed up the detection process.

4.4.3 Selection of Characteristic Scales

One of the considerations in the project is to find a suitable characteristic scale for use in the detection of a blood-pressure pulse. A characteristic scale is a range of wavelet transform scales employed in the detection process, e.g. scale $j = 3$ to $j = 5$ and scale $j = 2$ to $j = 4$.

In one of the experimental tests, around 20 different people had their blood-pressure measurement signals taken from age group around 24 years old students. Blood-pressure pulse trains were collected for three different test condition: a subject at rest, a subject walking with his/her arms still and a subject walking with his/her arms swinging. From the collected pulse trains, samples were extracted corresponding to three different regions of the pulse train:

- Systolic-region pulses
- Middle-region pulses
- Diastolic-region pulses

The objective of this extraction is to study the energy distribution of the various pulses under different situations, as well as under different cuff pressures. This is an important study because the algorithm to be designed can ignore these situational factors by using fixed characteristic scales. It would work if, and only if, the energy distribution of the pulses does not vary too much under different conditions. Otherwise, the algorithm has to take the different conditions into account.

Table 3. Frequency ranges of the pulses

	<i>stationary (Hz)</i>	<i>walking (Hz)</i>	<i>Swinging (Hz)</i>
diastolic	10 - 150	10 - 165	10 - 100
systolic	10 - 120	10 - 110	10 - 120
middle	10 - 180	10 - 175	10 - 160
overall	10 - 150	10 - 160	10 - 150

From the results shown in Table 3, two conclusions were made [6]:

1. Energy distribution of the individual blood-pressure pulse will not be affected much by the situational facts like a subject at rest, a subject walking with his/her arms still and a subject walking with his/her arms swinging.
2. The energy distribution, however, did depend on the cuff pressure acting upon it.

The range of the upper limit ($f_{cut-off}$) varies from 110 Hz to 180 Hz when the cuff pressure changes. This is an important factor to consider for the detection of blood-pressure pulses. This range of frequencies crosses the boundaries of scale 4 and scale 5. Hence, in order to cover all the possible ranges, both scale $j = 2$ to $j = 4$ and scale $j = 3$ to $j = 5$ must be implemented into the algorithm to effectively detect the blood-pressure pulses.

4.4.4 Pulse-Detection Range Boundary (PDB)

The minimum pulse rate of a subject is greater than or equal to 40 pulses per minute (*ppm*), and the maximum pulse rate of a subject is lesser or equal to 190 *ppm* as described in [44]. Hence, we set a detection boundary in the algorithm

from 30 *ppm* to 200 *ppm* as a buffer. These values, when translated to sample positions, become the prediction boundary for pulse detection. It is the worst possible case for the existence of a pulse. Furthermore this boundary is adjusted adaptively during the pulse-detection process, as discussed later.

For example, in this project the sampling frequency used when recording the Korotkoff sound was 5500 *Hz*.

For a minimum heart rate at 30 *ppm*, the period between the occurrences of consecutive Korotkoff sounds is

$$= \frac{60}{30} \times 5500 \text{ Hz} = 11000 \text{ sample points}$$

For a maximum heart rate at 200 *ppm*, the period between the occurrences of consecutive Korotkoff sounds is

$$= \frac{60}{200} \times 5500 \text{ Hz} = 1650 \text{ sample points}$$

Then after a pulse is detected the next pulse would occur after 1650 sample points from the current position but before 11000 sample points from current position.

The above are the initial pulse detection boundaries (*PDB*), i.e. $1650 < B_i < 11000$ sample points, where B_i is the initial window of occurrence of the next Korotkoff pulse from the current position. After three or more pulses are detected, we can use the pulse periods to adaptively update these boundaries. The algorithm used is

The next PDB from $(P_n + T_a - 2S_a)$ to $(P_n + T_a + 2S_a)$

where,

P_n = Position of the last detected pulse;

T_a = Average periods of the detected pulses;

S_a = Standard deviation of the detected pulses.

Here we are moving forward in sample positions. If we are moving backward from the mid-point to the front of the pulse train, then we have

The next PDB from $(P_n - T_a - 2S_a)$ to $(P_n - T_a + 2S_a)$

4.4.5 Selection of Window Size for Pulse Detection (PDW)

Our observations show that the average Korotkoff blood-pressure pulse duration with sampling frequency of 5500 Hz in wavelet transform lasts for 150 sample points (148 is the actual value) with standard deviation of 50 points (49 is the actual value). Hence a window size of 200 sample points was chosen for the detection process. This is to ensure that the window size is large enough to cover the blood-pressure pulses, yet as small as possible, to minimize noise interference during detection.

4.4.6 Pulse Detection

After the initial processing and peak detection (Section 4.3), we are ready to step up the next stage to detect each Korotkoff pulse from the blood-pressure waveform. Wavelet transform of blood-pressure waveform will be used in the pulse-detection process.

Initially the window size (*PDW*) of 200 sample points is shifted throughout the pulse-detection boundary (*PDB*) (as discussed in Section 4.4.4), until the first blood-pressure pulse is detected. The detection range for the first pulse detection is set at 11000 samples (as seen in Section 4.4.4) from the initial central peak location. Since this is the first detection, we must use the largest possible boundary. The detection starting point is obtained from the peak detection algorithm which is described in Section 4.3. From the peak we work towards the beginning of the waveform. Subsequently, we start from the same peak and work towards the end point of the waveform detecting the Korotkoff pulses as we go along.

After the first pulse from the central peak is detected, based on the lower limit of the blood-pressure pulse period, no pulse will occur for the next 1650 sample points. There is no point in scanning for pulses within that area. Besides that, based on the upper limit of the blood-pressure pulse period, the next pulse will be within 11000 sample points from the current (first) detected pulse. Hence the detection range will be 1650 to 11000 sample points from the current detected pulse location. This will help to eliminate noise that is outside this region and make the detection process faster.

The window size (*PDW*) of 200 sample points is shifted throughout the detection range until the next/second pulse is detected. If no pulse is detected within this range, invalid code will be saved and the function will be repeated first with a lower threshold as described below.

Now, if no pulse is detected within the stipulated window, there are basically three possible reasons:

1. There is no pulse within this pulse-detection boundary (*PDB*);
2. The pulse amplitude was less than the current threshold value;
3. The noise masked out the signal that make the SNR very small.

From these three conditions, the second reason is the most probable and thus the need for an adaptive threshold (refer to Section 4.2). If after lowering the threshold, the algorithm still cannot detect any pulse within the pulse boundary, it can be concluded that there is no blood-pressure pulse or the SNR is so small that the signal is totally masked out by the noise.

Thereafter we will first expand the *PDB* range from $\pm 2 S_a$ to $\pm 3 S_a$. However if still no pulse is detected in the longer boundaries we then continue with the pulse detection process in the next possible location. The determination of the next possible location is done by extending from the current pulse detection boundaries (*PDB*).

Here we assumed that the current pulse detection boundaries (*PDB*) contains a Korotkoff pulse although it is not detected. Then we just open the next pulse

detection boundaries (PDB) without a change in size and continue with the detection process.

For example, if we cannot detect a pulse in PDB from $(P_n + T_a - 3S_a)$ to $(P_n + T_a + 3S_a)$ boundaries, then we try to detect a pulse in PDB from $(P_n + 2T_a - 3S_a)$ to $(P_n + 2T_a + 3S_a)$ boundaries. This process is repeated again if no pulse is detected in the second detection boundaries. Hence we use PDB from $(P_n + 3T_a - 3S_a)$ to $(P_n + 3T_a + 3S_a)$. If still no pulse is detected within the three consecutive detection boundaries, two things may have happened:

1. Present pulse is a false detection, hence, subsequent pulse is missed;
2. Present pulse is the end of blood-pressure pulse.

The second reason is not difficult to determine. The criteria for deciding whether the present pulse is the end of blood-pressure pulse train or it is a false detection, is the number of blood-pressure pulses that have been detected so far and/or the location of the present sample points. If the number of detected blood-pressure pulses already exceed 20 or if the current sample point is less than 0 sample points (at the front of the pulse train) or more than 198000 sample points (at the tail of the pulse train), we can stop the detection process.

When no pulse can be further detected and the above conditions are not valid, the detection will terminate. An invalid code indicating failure of detection process will be saved for further use. It must be stated that the probability of failing to detect three consecutive pulses is negligible.

Because the algorithm calculates the mean of the pulse duration, having more pulses detected means that the prediction value becomes more accurate. Therefore the higher the number of pulses being detected, the more accurate the mean pulse duration calculated. This accuracy allows certain irregularities in pulse duration, without affecting the detection range of the next pulse.

4.4.7 Overall Pulse-Detection Procedure

This section will describe the blood-pressure pulse detection process in a step-by-step algorithm. The approach of this section is also described in a flowchart in Appendix A. Before we start the pulse-detection procedure, of course we must determine the constant values for the detection, such as the detection boundary (PDB), detection window size (*PDW*), threshold value etc. In the following description unless otherwise stated, we are examining the blood-pressure pulse train in the forward direction from the mid-point to the end of the waveform. The procedure is similar when moving from the mid-point to the front of the pulse train.

Step 1. Determine the maximum and minimum points and their location within the detection window for the different scales $j = 2$ to $j = 4$ of the quadratic-spline wavelet transform of the blood-pressure signal;

Step 2. Check whether maximum point has occurred after the minimum point for all the scales and whether their amplitude is increasing while the scale is increasing;

Step 3. Check whether their maximum and minimum value is higher than the threshold value that we had determined before (as described in Section 4.2) and their locations are not at the edge of the detection window (PDW);

Step 4. If Steps 2 and 3 are fulfilled, then we can conclude that a blood-pressure pulse has been detected. Save the detected pulse position for the next stage of the blood-pressure detecting algorithm. Determine the next threshold from the amplitude value of current detected blood-pressure pulses (Section 4.2). Determine next detection boundaries from the current and previously detected pulse periods (Section 4.4.4). The new pulse-detection boundaries are taken as the average of the already detected pulse periods $\pm$ two standard deviations, i.e. from $(P_n + T_a - 2S_a)$ to $(P_n + T_a + 2S_a)$ as described in Section 4.4.4. If the current detected pulse is the first one detected, we will use the initial range (B_i) as calculated in Section 4.4.4;

Step 5. If Steps 2 and 3 are not fulfilled, the pulse detection window (PDW) is shifted 10 sample points (5500 Hz sampling frequency) within the pulse detection boundaries and the checking is done again from Step 1. Such consecutive detection window is used if a pulse is still not detected. We continue until we reached the end of the detection boundary (PDB);

Step 6. If no pulse is detected within the entire detection boundary, the threshold will be reduced to 78% and the above steps will be

repeated again from Step 1. The above process will be repeated once more if no pulse is detected. This is to enable the reduction of the threshold value by 40%;

Step 7. After the third time and still no pulse is detected, the threshold will be reset to the original value and the detection window pointer will be reset to the starting point of the detection range. Then Steps 1 to Step 6 are repeated again but for a different scale $j = 3$ to $j = 5$;

Step 8. If the process in Step 7 did not detect any pulse, the process will be repeated for the scale $j = 2$ to $j = 4$, but this time round it will check for the maximum point before the minimum point. The threshold value is reset to the original value, and the detection window pointer is reset to the starting point of the detection range. The steps taken will be similar to Step 1 to Step 6. Again, it is done 3 times with different thresholds, if necessary;

Step 9. After 9 times of iteration and still no pulse is detected, we expand the detection range to thrice the standard deviation of the durations obtained from previously detected blood-pressure pulses. We assume that maybe the current pulse is at the edge of PDB. Here PDB from $(P_n + T_a - 3S_a)$ to $(P_n + T_a + 3S_a)$. The threshold value is reset, the pointer is reset to the starting point of the detection range and the checking is done again, as from Step 1. For example, if the current boundaries is from sample points $n_1 = (1000 - 100)$ to $n_2 = (1000 + 100)$, then the new

detection range is from $n_1 = (1000 - 150)$ to

$n_1 = (1000 + 150)$. Initially we have 200 points detection range.

After expansion the detection range will be 300 points;

Step 10. Step 9 uses the original threshold for the pulse detection. If after expanding the detection boundaries or after reaching the limit of possible detection range, i.e. 11000 sample points from position of current pulse and we still cannot detect any new blood-pressure pulse, then we try to decrease the threshold once more. The threshold value is decreased, the pointer is reset to the starting point of detection range in Step 9, and the checking is done again with the reduced threshold, as from Step 1 to Step 8;

Step 11. If still no pulse is detected, it could be because the signal is too weak or can't be compare with the threshold, then we leave the current detection boundaries and proceed with detection process on the next possible detection locations. The next PDB is from $(P_n + 2T_a - 3S_a)$ to $(P_n + 2T_a + 3S_a)$ followed by $(P_n + 3T_a - 3S_a)$ to $(P_n + 3T_a + 3S_a)$ if still no pulse is detected. Finally if undetection of pulse happens three times consecutively then the detection algorithm will be terminated.

The reason for not combining the above functions together (e.g. scale $j = 2$ to $j = 4$, scale $j = 3$ to $j = 5$ and detection of minimum peak before maximum peak), or by using the "OR" function is because it will increase the probability of false detection.

Since 90% of the time, scale $j = 2$ to $j = 4$ minimum peak followed by maximum peak is good enough to detect the blood-pressure pulse, only a minority of time will detection fall in the other two criteria. Hence, it is reasonable to use scale $j = 2$ to $j = 4$ minimum peak followed by maximum peak as a top priority for detection.

4.5 Systolic and Diastolic Point Detection

After detecting all the Korotkoff pulses and saving their positions, the final step is to determine the systolic and the diastolic points location from these detected blood-pressure pulses. By knowing all the blood-pressure pulse positions, we can extract a new waveform with only the blood-pressure pulses and minimal artifacts. An example of the original blood-pressure waveform and the new waveform that was formed from the detected blood-pressure pulses is shown in Figure 8.

From the waveforms in Figures 5(a) and (b), we can conclude that the systolic point is determined by the first increase of amplitude of the blood-pressure pulse and the diastolic point is determined by the first increase in amplitude of blood-pressure pulses in the reverse order, i.e. when blood-pressure waveform is traversed from end point to the beginning. From Figure 5(c), the systolic point is determined by the first occurrence of blood-pressure pulse and the diastolic point is determined by the last occurrence of blood-pressure pulse.

Since the occurrence of waveforms corresponding to that of Figures 5(a) and (b) is much higher than the type of waveform corresponding to Figure 5(c), the algorithm that was used for the detection of the systolic and diastolic points should be based first on Figure 5(a) and (b). If systolic and diastolic points cannot

be detected by the said method, then the algorithm that was used to detect systolic and diastolic points will be changed to that based on Figure 5(c). If both algorithms fail to locate systolic or diastolic points, then an error would be indicated.

Figure 8. Original and detected-pulse waveform plot

When the arms moves, the main envelope of the blood-pressure pulses is still maintained, but the variation of amplitude between pulses may vary as much as 40%. Hence, the waveform will not look exactly like Figure 5(a). There will be no consistent increase in amplitude at the beginning of the waveform and no consistent decrease amplitude at the end region.

Because of the above problem, the detected blood-pressure waveform was first passed through a median filter to smooth out the envelope, so that the variation will be minimized before passing the blood-pressure waveform for detection. The detection algorithm based on Figures 5(a) and (b) is not too difficult, since we can find the highest change in slope in the envelope of the blood-pressure pulses. We will treat the pulse where the slope changes as the systolic point. For diastolic detection, the detection is scanned from the back of the waveform to front, using the same algorithm.

If the above detection algorithm fails to detect either the systolic or the diastolic points, then the respective points will be detected using another algorithm. From Figure 5(c), we see that this kind of waveform will have more or less constant amplitude, therefore a threshold of 50% of maximum amplitude of the original blood-pressure pulse is set. The first pulse will be treated as a systolic point, or the last point will be treated as diastolic point, but only when the amplitude of all blood-pressure pulses are above threshold and the variance of the peak values is less than 10% of the mean amplitude value.

4.6 Results

The method described above was used to detect each pulse from the blood-pressure waveform. Then the location of systolic and diastolic points were determined. The results of the method used to determine the locations of systolic and diastolic points are in time unit. They indicate the times in which the systolic and diastolic pressure occurs. They are not presented in mm Hg unit. We can change the result from time unit (s) to pressure unit (mm Hg), since the cuff pressure is correlated to the time. The cuff pressure can be read from a pressure transducer attach to the cuff.

In order to validate the result from the proposed method, we try to compare our result of blood-pressure measurement with the blood-pressure measurement done by doctors or nurses using the same Korotkoff sound sources. The measurement of blood pressure is done by listening to the Korotkoff sound. Thus, the result can be subjective, since every person has different ability (sensitivity) in hearing. Therefore, more than one doctor or nurse are used to determine the location of systolic and diastolic points of the blood pressure of every subject that we are trying to measure.

In this case, we have 6 persons, 2 doctors and 4 nurses, to perform the blood-pressure measurement by listening to the Korotkoff sounds. Their age are varying from 25 years old to 35 years old. Then we calculate the average and the deviation of their results. Finally, we compare the values from the above measurements with our results from the proposed algorithm.

The following 3 tables (Table 4, Table 5 and Table 6) show the comparison of blood-pressure measurement between the values from the doctors/nurses and the outcome from our proposed algorithm that we run using Matlab software.

The numbers in the tables (Table 4, Table 5, and Table 6) indicate the time unit where the systolic and diastolic points occurred. The Systolic and Diastolic columns show the average time of occurrence with two standard deviation from the results we get from the doctors and nurses. The Detected by Algorithm column is the outcome value from our algorithm. From those tables, in the Detected by Algorithm column, we can see that in the several rows have check mark (✓). The check mark sign (✓) shows that the value in the Detected by algorithm column is in the range of systolic or diastolic column value.

Table 4 shows the results for cases where the measurement were done on 17 stationary subjects. Table 4 divide into 3 sections, the first 5 rows shows that the algorithm result, both of systolic and diastolic values, are in the range of results from doctors and nurses. In this case our algorithm work perfectly like our expectation. The next 9 rows shows that our algorithm can detect 7 systolic value and 2 diastolic value. The last 3 rows shows that our algorithm results are different from the doctors and nurses results, especially the diastolic value.

We can conclude that our algorithm almost success to detect all the blood-pressure waveform from stationary subject. However, few of them we cannot determine the systolic and/or diastolic points, this may happen when the waveform is corrupted with another sound.

Table 4. Results on stationary subject

Subject	Systolic		Detected by algorithm	Diastolic		Detected by algorithm
by_s	10.00	± 5.73	10.5 ✓	30.71	± 4.80	32.13 ✓
by2_s	7.14	± 4.38	11.49 ✓	24.57	± 9.30	32.18 ✓
kimhck_s	12.33	± 9.42	11.36 ✓	23.80	± 13.98	23.87 ✓
lck2_s	6.43	± 5.43	11.57 ✓	25.14	± 7.62	22 ✓
lowhm_s	22.14	± 29.73	2.75 ✓	32.50	± 18.36	32.33 ✓
jinsan_s	2.17	± 2.94	1.52 ✓	21.50	± 1.65	34.48
os_s	5.40	± 9.39	0.42 ✓	11.50	± 3.00	22.83
kimhk2_s	10.57	± 2.37	8.87 ✓	19.67	± 3.63	28.98
lck_s	3.86	± 6.81	0.13 ✓	22.14	± 3.63	29.69
ricky2_s	5.43	± 3.81	8.44 ✓	18.29	± 5.67	26.17
ricky_s	4.86	± 4.38	1.26 ✓	20.29	± 10.65	33.29
yklew_s	6.14	± 6.57	0.85 ✓	14.00	± 1.89	35.87
jnsan2_s	2.71	± 2.28	13.65	25.71	± 7.08	29.58 ✓
leekk_s	13.67	± 1.56	8.80	27.20	± 13.80	25.35 ✓
angeb_s	16.86	± 6.81	2.53	30.57	± 2.37	35.30
chgin2_s	13.14	± 6.81	4.67	23.29	± 8.79	35.90
os2_s	3.67	± 5.94	10.77	19.00	± 2.13	35.20

Table 5 shows the results for cases where the measurement were done on 13 subjects walking but with the arms held still. Different with the previous Table 4, Table 5 does not shows any no match for both systolic and diastolic points between the algorithm and the doctors and nurses result. Table 5 divide into 2 sections, the first 5 rows shows that the algorithm result, both of systolic and diastolic values, are in the range of the results from doctors and nurses. The next 8 rows shows that our algorithm can detect 3 systolic value and 5 diastolic value.

Table 5. Results on subject walking with arms still

Subject	Systolic		Detected by algorithm	Diastolic		Detected by algorithm
by_w	8.00	± 3.00	8.58 ✓	25.29	± 3.33	22.69 ✓
ken2_w	11.71	± 3.75	12.45 ✓	23.29	± 1.47	22.38 ✓
ken6_w	12.50	± 6.21	11.8 ✓	31.67	± 3.09	31.04 ✓
leekk_w	12.57	± 11.46	10.17 ✓	28.29	± 9.27	31.44 ✓
teykh_w	14.20	± 8.04	11.73 ✓	18.67	± 12.24	23.60 ✓
teykh2_w	7.17	± 3.51	7.82 ✓	12.40	± 1.65	27.39
wongcs_w	12.67	± 1.56	13.61 ✓	27.83	± 1.23	31.49
yeohos_w	17.71	± 6.63	13.96 ✓	31.43	± 1.59	33.76
ken1_w	16.67	± 1.56	12.74	32.17	± 1.23	32.59 ✓
lewyk3_w	15.33	± 7.02	4.59	28.83	± 12.51	24.73 ✓
lowhm_w	10.43	± 6.90	2.33	29.00	± 18.24	30.80 ✓
ken3_w	16.00	± 8.28	0.12	32.00	± 1.89	30.62 ✓
ken4_w	14.00	± 2.67	8.34	23.83	± 1.23	24.79 ✓

Table 6 indicates the results for cases where the 11 subjects were walking with the arms swinging freely. Like Table 4, this result divide into 3 sections. The first 3 rows shows that the algorithm result, both of systolic and diastolic values, are in the range of the results from doctors and nurses. In this case our algorithm work perfectly like our expectation. The next 5 rows shows that our algorithm can detect 3 systolic value and 2 diastolic value. The last 3 rows shows that our algorithm results are different from the doctors and nurses results.

Table 6. Results on subject walking with arms swing

Subject	Systolic		Detected by algorithm	Diastolic		Detected by algorithm
leekc1	10.33	± 7.74	4.89 ✓	30.33	± 12.24	31.11 ✓
leekk1	3.00	± 7.35	5.68 ✓	21.67	± 7.02	21.81 ✓
yeohos	10.83	± 9.18	9.35 ✓	32.00	± 13.29	29.46 ✓
by1	7.17	± 1.23	8.13 ✓	25.40	± 2.67	32.99
lewyk1	3.00	± 1.00	2.73 ✓	13.83	± 5.49	21.30
lowhm1	6.00	± 1.00	6.31 ✓	17.33	± 2.46	27.22
ken7	9.00	± 1.00	7.66	33.67	± 2.46	34.85 ✓
paul1	8.83	± 2.25	2.48	21.00	± 25.20	32.84 ✓
cgin	16.17	± 2.94	6.85	22.83	± 1.23	34.99
ken1	1.50	± 1.65	7.58	18.00	± 1.00	32.97
ricky1	4.00	± 1.00	6.16	27.00	± 3.30	32.77

Chapter 5

Conclusions & Recommendations

5.1 Conclusions

In this thesis, we present an algorithm for automatic blood-pressure measurement using the non invasive auscultatory method. The algorithm focuses on detecting Korotkoff blood-pressure pulses for non-stationary subjects. In the algorithm, the quadratic-spline wavelet transform method has been adopted to determine each pulse of the blood-pressure waveform. Finally the algorithm will determine the systolic and the diastolic points of the blood-pressure signal.

However, errors sometimes occur in the blood-pressure pulse-detection process, especially on non-stationary subject signals, because of the presence of strong ambient noise and motion artifacts. The first three blood-pressure pulses detected in the beginning are the most important. This is because the determination of the subsequent blood-pressure pulse locations depends on the blood-pressure pulses that have occurred earlier. If the first three blood-pressure pulses were wrongly detected, this will cause the subsequent signal detection to be in error. As more and more of the blood-pressure pulses were detected, this will make the determination of the subsequent blood-pressure pulse location faster and more accurate.

The disadvantage of the proposed method is the computational cost for the iterative procedure. However, the detection results in the previous tables (Table 4, Table 5, and Table 6) showed positive progress on the main objective to determine the systolic and the diastolic points of the blood-pressure signal. More research in this topic can sharpen the result in future.

5.2 Recommendations for future work

Although much progress has been achieved in our research, future work is needed to make the proposed algorithm more complete:

- Trying to re-code the algorithm by using another language instead of using Matlab software. The new program should be more flexible and more efficient on the use of the memory resource.
- The threshold value in our algorithm can be more precise if we have more subjects that can be used to carry out statistical study on them.
- More trained personnel (doctors/nurses) to help us with the manual determination of systolic and diastolic points. This can make the comparison more accurate and meaningful.
- Increase the number of subjects being observed and group the subjects, based on age, sex, conditions etc.

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1. Initialization of the constants.
2. Detection turn flag ≤ 1 .
3. Initialization of the variables.
4. Is it the first turn of the blood-pressure pulses detection?
5. Determine the first threshold and detection boundary.
6. Is it the second turn of the blood-pressure pulses detection?
7. Determine the next threshold and detection boundary from the detected blood-pressure pulses.
8. Is detection blood-pressure pulses flag $== 0$?
9. Determine the max and min amplitude values, including their positions, in the detection window at different scales 2 to 4.
10. Increase the detection flag by 1.
11. Is there a blood-pressure pulse condition fulfilled?
12. Blood-pressure pulse detected condition. (see the next flowchart, page 70).
13. Blood-pressure pulse not detected condition. (see the next flowchart, page 71).
14. Is the detection boundary position out of the blood-pressure waveform boundary?
15. Increase the detection blood-pressure pulses flag.

Blood Pressure Pulse Detected Condition

1. Increase the number of blood-pressure pulses counter.
2. Is it the first turn of the blood-pressure pulses detection?
3. Save the detected pulse location.
4. Determine the next threshold from previous detected pulse.
5. Is it the second turn of the blood-pressure pulses detection?
6. Save the detected pulse location.
7. Determine the next threshold from previous detected pulse.
8. Reset the variables.
9. Determine the next detection boundary from the previous detected pulse location.

Blood Pressure Pulse Not Detected Condition

1. Is the detection window exceeding the detection boundary?
2. Is the threshold counter more than 2?
3. Shift the detection window.
4. Is the step counter equal to 3?
5. Decrease the threshold value and increase the threshold counter by 1.
6. Is the detection boundary in the range of determined location?
7. Increase the step counter by 1, reset the threshold counter and the threshold value.
8. Increase the reduce counter by 1.
9. increase the expanding counter by 1, reset the threshold value, and determine the new detection boundary.
10. Is the reduce counter equal to 2.
11. Is the blood pressure detected counter less or equal to 3.
12. Determine the new detection boundary and decrease the threshold value.
13. Determine the next detection boundary.
14. Increase the missing detection counter by 1.
15. Reset the expanding counter.
16. Reset the step counter.
17. Reset the threshold counter.
18. Reset the shift counter.