Robust Estimation of Mean Arterial Pressure in Atrial Fibrillation using Oscillometry

By

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Author's Declaration

I hereby declare that I am the sole author of this thesis. This is a true copy of the thesis, including any required final revisions, as accepted by my examiners.

I understand that my thesis may be made electronically available to the public.

Milad Tannous

Abstract

Blood pressure measurement has been and continues to be one of the most important measurements in clinical practice and yet, it remains one of the most inaccurately performed. The use of oscillometric blood pressure measurement monitors has become common in hospitals, clinics and even homes. Typically, these monitors assume that the heartbeat rate remains stable, which is contrary to what happens in atrial fibrillation.

In this thesis, a new method that provides a more precise estimate of Mean Arterial Pressure (MAP) is proposed using a non-invasive oscillometric blood pressure monitor. The proposed method is based on calculating a ratio of peak amplitude to trough amplitude for every pulse, then identifying where the ratio first reaches a value of 2. The performance of the proposed method is assessed by comparing the accuracy and variability of the readings against reference monitors - first in healthy subjects, then in atrial fibrillation patients.

In healthy subjects and in atrial fibrillation patients, the proposed method achieved a performance accuracy that is well within the ANSI/AAMI SP10 protocol requirements of the reference monitors. The presence of atrial fibrillation diminished the performance of the reference monitor by increasing the variability of the reference readings. The proposed algorithm, on the other hand, performed better by achieving substantially lower variability in the readings than the reference device.

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List of Abbreviations

SBP: Systolic Blood Pressure

DBP: Diastolic Blood Pressure

MAP: Mean Arterial Pressure

PP: Pulse Pressure

mmHg: Millimeters of Mercury

ECG: Electrocardiogram

PPG: Photoplethysmograph

PTT: Pulse Transit Time

MAA: Maximum Amplitude Algorithm

OMW: Oscillometric Waveform

BHS: British Hypertension Society

ANSI: American National Standards Institute

AAMI: Association for the Advancement of Medical Instrumentation

ME: Mean Error

STD: Standard Deviation

STDE: Standard Deviation of Errors

MAE: Mean of Absolute Errors

USB: Universal Serial Bus

Chapter 1. Introduction

According to Statistics Canada, every seven minutes a person in Canada dies due to a stroke or a cardiovascular disease (Heart and stroke foundation, 2013). Cardiovascular diseases are related to the function of the cardiovascular system, made up of the heart and its blood vessels along with the veins, arteries and capillaries of the body and the brain. When the heart cannot pump as much blood as the body needs, the slow flow of blood creates pools that can result in blood clots. A stroke may occur if blood clots get dislodged and travel to the brain, blocking the blood flow through the arteries.

1.1 Blood Pressure

Blood pressure provides information related to the physiological state of a patient's cardiovascular system. It is the outward pressure exerted on the insides of the arterial walls by the circulating blood and is quantified in terms of millimetre changes in a mercury column (mmHg). Elevated blood pressure is referred to as hypertension, while the contrary is known as hypotension. The sensitivity of blood pressure to various external and internal physiological conditions augments its importance as a biomarker of health and stress (Webster, 1998).

Every cardiac cycle is divided into two phases: systole and diastole. Systole is the phase where the heart pumps the blood out from the left ventricle, where the pressure of the circulating blood on the arterial walls reaches its peak. That pressure value is referred to as Systolic Blood Pressure (SBP). Diastole is the phase where the ventricles relax and cause lower pressure against the arterial walls. The minimum pressure value is known as Diastolic Blood Pressure (DBP). Mean Arterial Pressure (MAP) is the average blood pressure for every heartbeat (Zheng et al., 2008). MAP reflects the fluctuating blood pressure better than SBP or DBP (Smulyan et al.,

2008) and also reflects the hemodynamics of tissue perfusion pressure in the organs (Nichols W., 1998). It is used to calculate the peripheral vascular resistance (Graf et al., 2010) and to calibrate central blood pressure waveforms (Vos et al., 2013). MAP is especially important when medication is prescribed to maintain an adequate perfusion blood pressure to damaged organs, such as in stroke patients (Razminia et al., 2004).

Figure 1.1 shows where SBP, DBP and MAP are located on a tracing of arterial blood pressure.

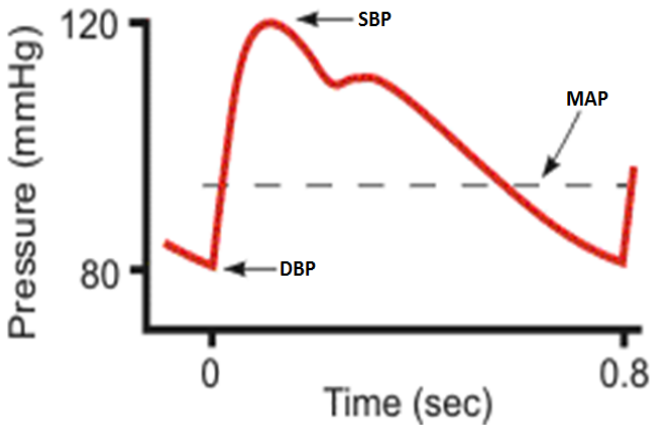


Figure 1.1 Arterial blood pressure trace over a cardiac cycle (adapted from Klabunde, 2007, with modifications).

Blood pressure can be measured invasively and non-invasively and the measurement procedure can be performed either manually or automatically. The invasive intra-arterial method is considered to be the most accurate method for blood pressure measurement. However, it requires the insertion of a catheter into an artery, surgically, by a highly trained operator. This procedure is accompanied by a risk of bleeding, infection or blood clotting (Turner, 2000).

Non-invasive methods are more commonly used as they are safer, faster and easier to perform. The most common non-invasive methods used in clinical practice are auscultation and oscillometry. The auscultatory method makes use of a sphygmomanometer and a stethoscope. Connected to the sphygmomanometer is a cuff that is wrapped around the patient's upper arm.

The pressure cuff is inflated above the expected SBP and then deflated at a steady rate, typically around 3 mmHg/sec (Pickering et al., 2005). During the deflation process, an observer would place the stethoscope on the brachial artery below the cuff and listen to the five Korotkoff sound phases. The sphygmomanometer reading when the Korotkoff sounds first become audible, Phase I, is marked by the observer as SBP and the reading corresponding to the onset of Phase V, where the Korotkoff sounds completely disappear, is marked as DBP.

Oscillometry is considered the most common automated method, especially in the doctor's office or in a home setting, due to its simplicity, speed, and robustness. As with the auscultatory method, a cuff is inflated above the expected SBP and then deflated steadily. Unlike the auscultatory method, rather than relying on the auscultations of the Korotkoff sounds, a pressure transducer in the cuff records the pressure pulsations of the blood in the arteries during the deflation stage. The recorded signal is analyzed and processed digitally to obtain MAP, and then to calculate SBP and DBP.

1.2 Atrial Fibrillation

Affecting 350,000 Canadians (Heart and Stroke foundation, 2013), 2.2 million Americans and 4.5 million Europeans (Fuster et al., 2006), atrial fibrillation is the most common arrhythmia in clinical practice (Stein et al., 1999; Jani et al., 2006; Corino et al., 2008; Lamb et al., 2010). It is an irregular atrial depolarization that causes an irregular ventricular rhythm (Stein et al., 1999). Normally, the chambers of the heart contract in a very organized way, so the heart can pump all the blood the body needs with minimal effort. In atrial fibrillation, the electrical impulse of the heart loses its regularity, causing the atria to contract rapidly in an irregular pattern. This causes the ventricles to beat abnormally, leading to an irregular pulse. If the heart cannot pump enough

blood, some blood pools and forms clots. If blood clots get dislodged and travel to the brain, the blood flow through the arteries could get blocked, causing a stroke.

1.3 Motivation

One in five Canadian adults (Wilkins et al., 2010) and one in three American adults (National Center for Health Statistics, 2013) have high blood pressure, also referred to as hypertension. Untreated hypertension may lead to other cardiovascular diseases and happens to be one of the most common causes of atrial fibrillation (Psaty et al., 1997). The risk of a stroke increases by up to five fold in the presence of hypertension or atrial fibrillation (Heart and Stroke Foundation, 2013). This is why blood pressure measurement has been and continues to be one of the most important measurements in clinical practice and yet, it remains one of the most inaccurately performed (Pickering et al., 2005).

Even though it is considered to be the method of choice for non-invasive blood pressure measurement in clinics and offices (Pickering et al., 2005), studies have shown that the auscultation method tends to underestimate SBP and overestimate DBP (Noaman and Abbas, 2007). Due to factors that include the white coat effect or handling errors, such as selecting an inappropriate cuff size or deflating the cuff too fast, such measurements are vulnerable to errors.

The use of automated oscillometric monitors has become common in homes and clinics but patients with abnormal heartbeat rhythms struggle with the accuracy of the results. The beat-to-beat variability of blood pressure increases in atrial fibrillation due to the rapid variations in the left ventricle's contractility and filling time (Jani et al., 2006), leading to erroneous estimations of MAP, and consequently SBP and DBP.

Researchers have argued against the use of oscillometric monitors when making therapeutic or medication-related decisions for patients with abnormal heartbeat rhythms due to the large error margin found in the blood pressure estimates(Anastas et al., 2008), claiming that no oscillometric monitor has been validated, in patients with atrial fibrillation, through any recognized protocol (Lamb et al., 2010).

MAP is a very important hemodynamic parameter and an accurate estimate is necessary to calibrate the arterial pulse waveforms used to calculate blood pressure. According to Vos et al. (2013), a precise measurement of MAP can be obtained only with the invasive intra-arterial approach, even in healthy subjects. MAP displayed by the non-invasive oscillometric monitor (WatchBP Office Monitor) investigated in their study was "too imprecise to be used for calibrations" (Vos et al., 2013).

The work presented in this thesis addresses the inaccuracy of clinical- and home-based automated blood pressure monitors when used by atrial fibrillation patients, who may or may not even be aware of its presence, by proposing a novel approach for MAP estimation that does not require any hardware additions or modifications.

1.4 Thesis Overview

The use of oscillometric blood pressure measurement devices has become common in hospitals, clinics and even homes. Such monitors utilize proprietary algorithms that vary from device to device, to estimate systolic and diastolic blood pressure. Unfortunately, the poor performance found in patients with atrial fibrillation shows that these algorithms depend on heartbeat rate remaining regular. This is contrary to what happens in atrial fibrillation, the most common type of arrhythmia. (Stein et al., 1999; Jani et al., 2006; Corino et al., 2008; Lamb et al.,

2010), where an irregular atrial depolarization of the heart causes an irregular heartbeat rhythm (Corino et al., 2008).

In this study, a new method for estimating the mean arterial pressure non-invasively is explored, using an oscillometric blood pressure monitor that makes use of a cuff for the upper arm. The cuff is inflated until the cuff pressure is above a certain value and then begins deflating at a slow steady rate as the cuff pressure signal gets recorded. Signal processing is performed off-line to obtain a filtered version of the cuff pressure signal that reflects the arterial blood pressure oscillations throughout the deflation process. The algorithm then locates the peaks and troughs on the arterial blood pressure oscillation waveform and a ratio is calculated based on the maximum and minimum peak pressure values for each pulse. The oscillometric pulse which corresponds to the mean arterial pressure value was consistently found to be the one closest to or equal to two. The suggested algorithm is tested by estimating the mean arterial pressure of subjects, with and without atrial fibrillation, and comparing the results to ones obtained using different approaches.

Chapter 2. Background

Blood pressure fluctuates significantly from beat to beat during arrhythmias, when the heart's cardiac rhythm becomes very irregular, increasing the chances of inter- and intra-observer errors (Sykes et al., 1990). There are no generally accepted guidelines when using the Korotkoff sounds method as blood pressure estimation becomes a guess at best, while automated devices significantly lose accuracy in the presence of atrial fibrillation (Stewart et al., 1995). This chapter provides background on blood pressure, with a particular focus on Mean Arterial Pressure (MAP), and the methods used to measure it. It also describes the specific type of cardiac abnormality or arrhythmia, known as atrial fibrillation, and the common blood pressure measurement techniques and the problems that arise during blood pressure measurement in patients with atrial fibrillation.

2.1 Blood Pressure

Every heartbeat initiates a cardiac cycle that originates at the sino-atrial node, the heart's natural pacemaker. An electric wave travels through the atria before passing through the atrio-ventricular node, where it pauses before finally reaching the ventricles. The electric wave causes the ventricles to contract, leading to a squeezing motion by the heart that pumps the blood out of the heart to the rest of the body.

A mechanical pulse is generated in the heart with the expansion and contraction of the arterial walls due to the pumping of the blood from the heart, and therefore the pulse rate is also referred to as heartbeat rate as it is synchronous with the beating mechanism of the heart. The pulse rate is a count of the number of contractions or heartbeats occurring over a time interval, typically a minute. Pulse rate fluctuations may reflect changes in the patient's physiological, as

well as emotional, state. Reduced variations in the pulse rate, also known as heart rate variability, have been shown to be associated with cardiovascular conditions such as congestive heart failure (Abildstrom et al., 2003).

Every cardiac cycle is divided into two phases: systole and diastole. As seen in Figure 2.1, Systole is the phase at the beginning of each cardiac cycle, as the heart pumps the blood out from the left ventricle, where the pressure of the circulating blood on the arterial walls reaches its peak. That pressure value is referred to as Systolic Blood Pressure (SBP). Diastole, on the other hand, is the remainder of the cardiac cycle, where the ventricles relax and blood returns to fill the heart as it prepares for the next beat. Blood flow is slower during diastole and therefore results in lower pressure against the arterial walls. The minimum pressure value is known as Diastolic Blood Pressure (DBP) (Mohrman and Heller, 2010).

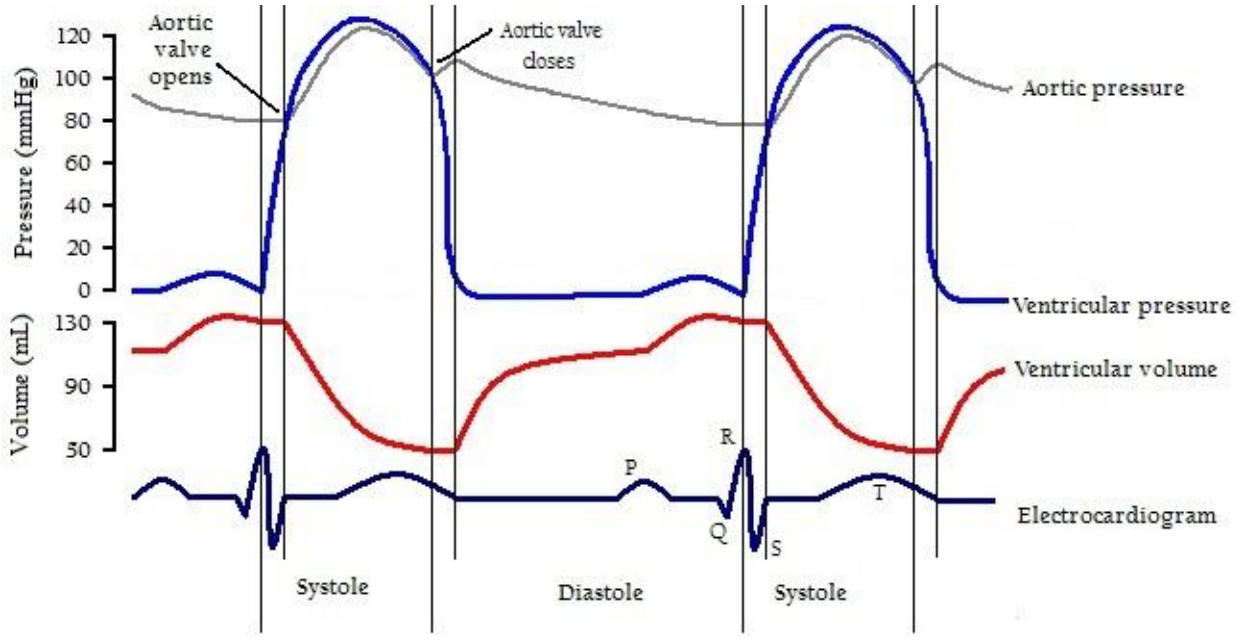


Figure 2.1 The systole and diastole phases in a cardiac cycle (adapted from Chang, 2011, with modifications)

2.2 Mean Arterial Pressure

Mean Arterial Pressure (MAP) is the average arterial blood pressure over an entire cardiac cycle and is determined by the flow equation (Pocock and Richards, 1999):

$$Q = P \div R \qquad \text{Equation 2.1}$$

where Q is the flow, P is the pressure and R is the resistance to flow. Since the cardiac output is the amount of blood pumped over a period of time (flow) and the total peripheral resistance is the combined vascular resistance of the circulatory system (resistance to flow), this equation can be rearranged into the following form:

$$\text{MAP} = \text{Cardiac Output x Total Peripheral Resistance} \qquad \text{Equation 2.2}$$

MAP is an important cardiovascular parameter because it is the average pressure at which the blood is pumped to the body's organs (Mohrman and Heller, 2010). From the organs' point of view, MAP is the perfusion pressure (Pocock and Richards, 1999).

MAP has been used as a predictor of cardiovascular diseases (Sesso et al., 2000) such as Ischemia (Zheng et al., 2008). According to Edwards Lifesciences, a medical equipment company and a global leader in hemodynamic monitoring and artificial heart valves (Stynes, 2013), the normal MAP range is 70 - 105 mmHg (Edwards Lifesciences Co., 2012). Inadequate blood flow, due to MAP being below 65 mmHg (Polanco and Pinsky, 2006), can inflict ischemic damage to the body's organs, thus reducing the affected organ's performance. Poor blood flow to the brain is associated with lethargy and drowsiness and in some cases may induce a coma, whereas poor blood flow to the kidneys leads to numerous metabolic consequences (Walker et al., 1990).

2.3 Calculating Mean Arterial Pressure

Different formulas have been suggested in the literature to calculate MAP since commonly used non-invasive blood pressure monitors display SBP and DBP, and in some cases the heart rate, but not MAP.

2.3.1 33% Formula

If the arterial pressure waveform was perfectly sinusoidal, MAP would simply be the arithmetic average of SBP and DBP. However, that is not the case. MAP is a time-weighted average (Meaney et al., 2000) and since the cardiac cycle, at normal heart rates, is approximately one-third systole and two-thirds diastole, MAP is less than the arithmetic average and is closer to DBP than SBP.

For decades, MAP has been calculated from measured SBP and DBP simply by adding a third of SBP and two-thirds of DBP (Geddes et al., 1982).

$$\mathbf{MAP} = \frac{\mathbf{1}}{\mathbf{3}}\ \mathbf{SBP} + \frac{\mathbf{2}}{\mathbf{3}}\ \mathbf{DBP} \qquad \text{Equation 2.3}$$

In other words,

$$\mathbf{MAP_{33\%}} = \mathbf{DBP} + \frac{\mathbf{1}}{\mathbf{3}}\ \mathbf{PP} = \mathbf{DBP} + \mathbf{33\%\ PP} \qquad \text{Equation 2.4}$$

where PP (Pulse Pressure) is the difference between SBP and DBP (i.e., PP = SBP - DBP).

2.3.2 40% Formula

In 2000, Equation 2.4 was described as having "substantial limitations" (Meaney et al., 2000). Its validity was investigated by calculating MAP from SBP and DBP recorded by

auscultation and comparing it to invasively-measured MAP (Bos et al., 2007). The results showed that adding one-third of PP to DBP underestimates MAP, and the recommendation was to replace the one-third coefficient (i.e., 33%) by a coefficient of 0.4 (i.e., 40%) to compensate for the underestimation. This alternative 0.4 coefficient was further tested in another study (Verrij et al., 2008) where two MAP values were calculated, one using the one-third coefficient and the other using the 0.4 coefficient, from SBP and DBP recorded by auscultation. Applanation tonometry, calibrated with auscultatory measurements, was used to trace the brachial arterial pressure waveform and estimate the reference MAP in this study. The results showed that MAP calculated using the one-third coefficient was lower than the reference MAP and the authors concluded that the alternative formula, that uses the 0.4 coefficient, could be used to overcome this underestimation.

$$\mathbf{MAP_{40\%} = DBP + 0.4\,PP} \qquad \text{Equation 2.5}$$

2.3.3 HR-Dependant Formula

Both Equations 2.4 and 2.5 fail to take into consideration that an increase in heart rate (HR) reduces the durations of systole and diastole asymmetrically. In 2004, a new formula for calculating MAP that incorporates HR was investigated:

$$\mathbf{MAP_{HR} = DBP + (0.33 + 0.0012\,HR)PP} \qquad \text{Equation 2.6}$$

where HR is the Heart Rate (Razminia et al., 2004). MAP values were calculated, using both Equation 2.4 and the newly proposed Equation 2.6, and were compared with the reference MAP that was obtained by computing the area under an invasively-recorded pressure-time curve. The novelty of this study is that multiple measurements were recorded, for each of the 12 subjects, at several different heart rates ranging from 50 to 125 beats per minute.

Table 2.1 summarizes the results of the study, showing the mean ± standard deviation of reference MAP, MAP calculated using Equation 2.4, MAP calculated using Equation 2.6, difference errors between MAP calculated using Equation 2.4 and the reference MAP, and difference errors between MAP calculated using Equation 2.6 and the reference MAP across the different HR ranges as well as the entire range, where the differences are expressed as the Mean of the Errors (ME) ± STandard Deviation of the Errors (STDE). Based on those results, the authors conclude that Equation 2.6 is more suitable to calculate MAP because it adjusts for the effects of increased HR on systole and diastole.

HR Range (beat/minute)	Reference MAP (mmHg)	Calculated MAP (mmHg)		ME ± STDE (mmHg)	
		$MAP_{33\%}$	MAP_{HR}	$MAP_{33\%}$	MAP_{HR}
50 - 59	114 ± 9	110 ± 9	116 ± 9	- 4.00 ± 1.03	1.12 ± 1.25
60 - 69	119 ± 17	114 ± 16	121 ± 18	- 5.27 ± 2.23	1.41 ± 2.41
70 - 79	119 ± 21	112 ± 19	118 ± 21	- 6.41 ± 2.71	- 0.25 ± 1.50
80 - 89	122 ± 21	115 ± 19	122 ± 21	- 6.59 ± 3.12	- 0.20 ± 1.07
90 - 99	124 ± 23	117 ± 20	124 ± 23	- 7.22 ± 3.65	- 0.23 ± 1.15
100 - 109	115 ± 18	109 ± 16	114 ± 18	- 6.08 ± 3.04	- 0.42 ± 0.64
110 - 119	118 ± 16	112 ± 14	118 ± 16	- 6.12 ± 3.46	- 0.13 ± 1.65
120 - 129	116 ± 20	110 ± 19	116 ± 20	-5.56 ± 1.68	- 0.05 ± 0.76
Entire Range	119.5 ± 19.4	113.3 ± 17.3	119.5 ± 19.7	- 6.22 ± 2.98	0.07 ± 1.57

Table 2.1 Summary of the results of the study by Razminia et al. showing the difference between calculating MAP using Equations 2.4 and 2.6, and the difference errors between the reference MAP and MAP calculated using each equation - expressed as the Mean of the Errors (ME) ± STandard Deviation of the Errors (STDE) . The reference MAP and both calculated MAP values are expressed as the mean ± standard deviation (adapted from Razminia et al., 2004).

2.4 Atrial Fibrillation

Atrial fibrillation, the most common arrhythmia in clinical practice, is an irregular atrial depolarization that causes an irregular, but not completely random, ventricular rhythm (Stein et al., 1999). Normally, the chambers of the heart contract in a very organized way, so the heart can pump all the blood the body needs with minimal effort. The electrical impulse that signals the

heart to contract is initiated in the sino-atrial node, the heart's natural pacemaker. The signal then leaves the sino-atrial node and travels through the atria before passing through the atrio-ventricular node, and finally, through the ventricles. In atrial fibrillation, the electrical impulse of the heart loses its regularity, causing the atria to contract rapidly in an irregular pattern. This causes the ventricles to beat abnormally, leading to an irregular pulse. As a result, the heart cannot pump as much blood as the body needs. The slow flow of blood creates pools that can result in blood clots. If blood clots get dislodged and travel to the brain, the blood flow through the arteries could get blocked, causing a stroke. According to the Heart and Stroke foundation (Heart & Stroke foundation website, 2013), people with atrial fibrillation have a three to five times higher risk of stroke than those without it.

Atrial fibrillation can occur in two forms: paroxysmal atrial fibrillation, also known as intermittent atrial fibrillation, which is characterized by episodes that occur with varying frequency and last for a variable period of time before spontaneously stopping; and chronic or persistent atrial fibrillation, which is sustained and does not usually stop spontaneously.

A person with atrial fibrillation may experience palpitations (sensation of feeling the heartbeat), dizziness, shortness of breath, fatigue, fainting or a combination. On the other hand, some people may not even be aware of it, especially if it is the sustained type. The presence of atrial fibrillation can be confirmed using an Electrocardiogram (ECG) reading, where the absence of the P waves from the PQRST complex, as seen in Figure 2.2, and the irregular ventricular rate correspond to the presence of atrial fibrillation.

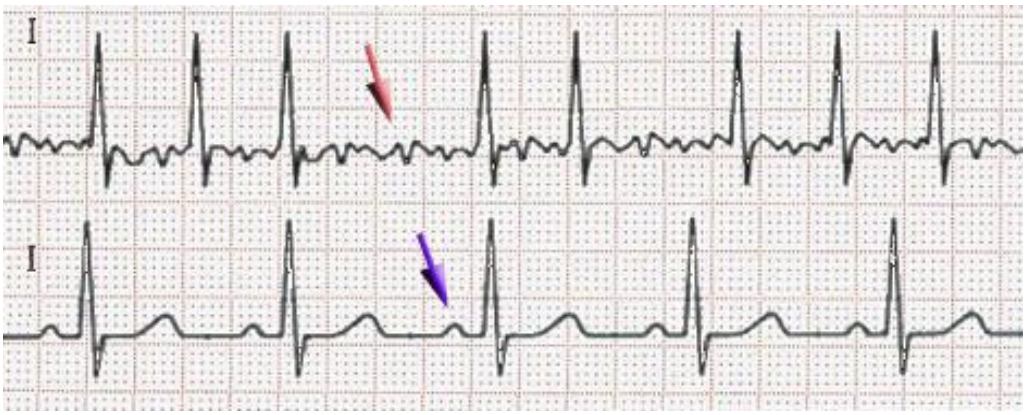


Figure 2.2 Top: ECG recording of a patient with atrial fibrillation. Red arrow points to where the P peak would be found in a healthy subject. Bottom: A typical ECG recording of a healthy subject. Blue arrow indicates presence of P waves (adapted from Heuser, 2005).

2.5 Blood Pressure Measurement

Blood Pressure measurement has been and continues to be one of the most important measurements in clinical practice and yet, it remains one of the most inaccurately performed. SBP and DBP, as well as MAP, may be measured in the clinic or the home, and can help predict vascular disease (Sesso et al., 2000). In a clinical setting, the current gold standard for non-invasive blood pressure measurement is auscultatory, using a mercury sphygmomanometer and a trained observer to apply the Korotkoff technique (Yarows and Qian, 2001). However, this procedure may lead to failure in detecting hypertension or misclassifications of healthy individuals as hypertensive. This is why it is important to be able to detect small changes in blood pressure accurately. Aside from the equipments' capabilities, inaccuracies occur as a direct result of the inherent variability of blood pressure. The white coat effect is a cause of error, too, as the blood pressure tends to increase in the presence of the physician, leading to an overestimation of SBP and DBP (Mion and Pierin, 1998; Rodriguez et al., 2010).

2.6 Invasive Blood Pressure Measurement

Invasive blood pressure monitoring gives the most accurate beat-to-beat information and is considered the gold standard of blood pressure measurement. This method is used in hospitals, and mainly in surgery rooms and intensive care units, for continuous monitoring of the patient's physiological state, when short-term blood pressure fluctuations are expected. Another advantage is its robustness in the face of factors that would reduce the accuracy of non-invasive blood pressure measurement methods, such as obesity or the presence of arrhythmias.

This method involves the manual insertion of a catheter, connected to a pressure transducer, into an artery, typically by a highly trained clinician (Webster and John, 1998). However, the high accuracy of this method comes at the cost of risking bleeding, infections and other complications such as thrombosis (Turner and Karen, 2000).

2.7 Tonometric Method

Tonometry is based on compressing an artery against a bone. Compressing the artery creates a flat-surface segment where the arterial pressure exerts force perpendicularly to the wall. A pressure transducer, placed directly on the skin over the artery, senses the intra-arterial forces and produces arterial pressure waveforms (Drzewiecki et al., 1983).

By placing a sensor above the radial artery, radial artery tonometry is able to provide a continuous waveform representing arterial pressure at that location. In 1996, a study was conducted with the purpose of evaluating radial artery tonometry in patients with cardiac diseases undergoing surgical procedures (Weiss et al., 1996). For the twenty-two high-risk surgical patients, the arterial pressure obtained by means of radial artery tonometry was compared with invasive radial artery pressure recordings. The authors concluded that radial

tonometry could replace invasive measurements in awake patients but should not be used as an alternative to invasive monitoring during high-risk surgical procedures. Siegel et al. (1994) reported in another study, that radial artery tonometry suffers from low accuracy and the "limited ability to detect significant changes more rapidly than with oscillometry alone".

In applanation tonometry, a single transducer is held manually over the radial artery to record the pressure waveform while systolic and diastolic pressures are measured from the brachial artery. This is used to estimate central aortic pressure.

Applying applanation tonometry at all arterial sites is not possible. It requires a stiff structure on which the artery wall can be flattened and a lean skin to avoid attenuating the pressure pulse signal. Applanation tonometry is also often inaccurate in obese patients. On the other hand, lean subjects produce better waveforms that can be obtained at the radial artery with ease (Nichols and O'Rourke, 1998).

Applying tonometry has its drawbacks. For example, the signal obtained is very position-sensitive because the transducer needs to be placed directly over the center of the artery. Another point of concern is that the equipment needs to be calibrated because tonometry only provides a waveform which in turn needs to be calibrated using other measurements.

2.8 Auscultatory Method

Auscultation remains the method of choice for non-invasive blood pressure measurement. A trained observer equipped with a stethoscope along with a pressure cuff and mercury sphygmomanometer are required. Due to environmental concerns, mercury sphygmomanometers are being used less nowadays, and are being replaced by aneroid devices. Other hybrid

alternatives use electronic transducers rather than mercury (Watson and Lip, 2006; Ward and Langton, 2007).

On the other hand, in mercury sphygmomanometers there are fewer parameters that could go wrong and the simplicity of the design means that different brands of such devices have identical levels of accuracy (Pickering et al., 2005).

Aneroid sphygmomanometers are devices used to record pressure using a system of metal bellows and levers that record the pressure on a cylindrical scale. The metal bellows expand as the cuff pressure increases, while the levers move in reaction and record the pressure. These devices are prone to losing their stability over time, especially if handled improperly. Manometers mounted on walls are less susceptible to motion artefacts and are therefore, inherently, more accurate than the mobile ones (Pickering et al., 2005). According to a number of experiments, the accuracy of such devices varies greatly from one brand to another. A few surveys, conducted in hospitals, demonstrated that aneroid devices, at the time of the experiments, have significant inaccuracies starting at 1% (Yarows and Qian, 2001) and reaching up to 44% (Mion and Pierin, 1998).

Some devices, known as hybrid sphygmomanometers, combine features from electronic and auscultatory devices, where an electronic pressure gauge replaces the mercury column. Such devices are operated by an observer, equipped with a stethoscope to listen for the Korotkoff sounds, in the same manner as other sphygmomanometers. Some devices display the cuff pressure as a simulated mercury column and others as a digital reading. Some have a built-in pause button that allows the observer to pause the reading at systolic and diastolic blood pressure

levels when encountering Phase I and Phase V of the Korotkoff sounds, respectively (Pickering et al., 2005).

2.8.1 Procedure

The cuff is placed around the upper left arm, with the stethoscope placed below the cuff, directly above the brachial artery. The cuff is then inflated until the Korotkoff sounds become audible and continues until they become inaudible again. Next, the cuff is deflated at a constant rate until the observer begins hearing the Korotkoff sounds, all through the five phases, until the sounds disappear. The five phases are classified as follows:

Phase I Appearance of clear tapping sounds.

Phase II Sounds become softer and longer.

Phase III Sounds become crisper and louder.

Phase IV Sounds become muffled and softer.

Phase V Sounds disappear completely. (Pickering et al., 2005)

The cuff pressure when the sounds become audible, at the starting point of Phase I (point a in Figure 2.3), corresponds to SBP, while the pressure when the sounds disappear, Phase V (point b in Figure 2.3), corresponds to DBP (Anastas, Jimerson, and Garolis, 2008).

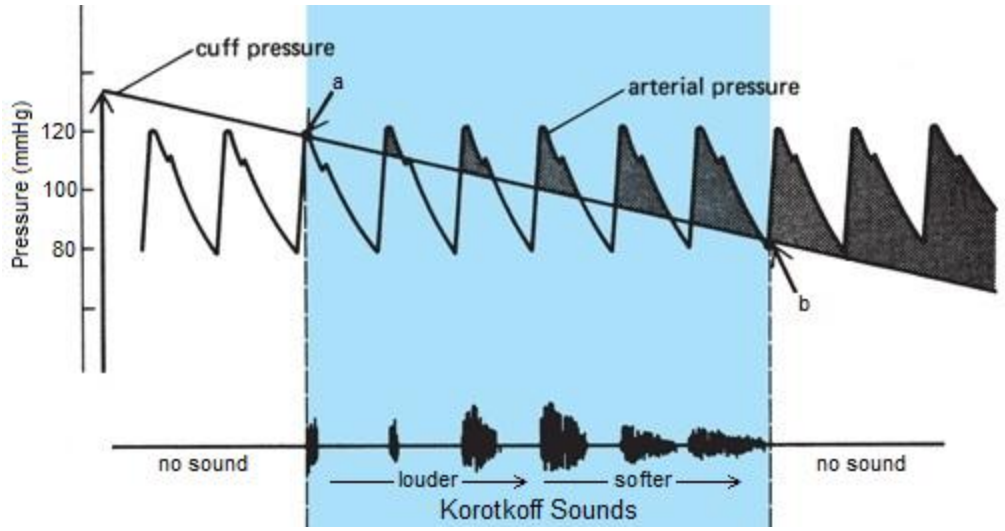


Figure 2.3 Using the Korotkoff sounds to determine SBP and DBP by auscultation. The cuff pressure at point a is SBP, while the cuff pressure at point b is DBP (adapted from Mohrman and Heller, 2010, with modifcation).

2.8.2 Challenges

Older patients may exhibit an auscultatory gap, where the Korotkoff sounds may become inaudible between the SBP and DBP points. This occurs due to the fluctuations in the intra-arterial pressure and is most likely to occur in patients with target organ damage. However, this gap can be eliminated by positioning the patient's arm overhead for 30 seconds before proceeding with the measurement, in order to improve inflow and enhance the Korotkoff sounds (Pickering et al., 2005).

In atrial fibrillation, as blood pressure fluctuates, the amplitude and pitch of the Korotkoff sounds changes. This increases the observer error due to the varying amplitude of the sounds. SBP tends to be underestimated because the observer might miss the first few beats of Phase I of the Korotkoff sounds, while DBP tends to be overestimated when the observer misinterprets the missed beats in Phase IV as the beginning of Phase V.

Kaliujnaya and Kalyuzhny (2005) argued that the Korotkoff method and all of its modifications can only provide a rough estimate of blood pressure in atrial fibrillation. They conducted a study introducing a new modified auscultative method of non-invasive blood pressure measurement that, according to the authors, demonstrated equal accuracy when compared to the direct invasive method, specifically in atrial fibrillation. This method is recommended when dealing with arrhythmias and it is based on step deflation, rather than a continuous steady rate, with ten seconds for each step, as the observer counts the number of Korotkoff sounds during each step of cuff pressure.

2.9 Oscillometric Method

The use of automated electronic monitors has become common in hospitals and clinics, with oscillometry being the most popular technique for such monitors (Sorvoja and Myllyla, 2006). Unlike the auscultation technique, many oscillometric monitors determine MAP first and then calculate SBP and DBP using empirically derived parameters that vary from manufacturer to manufacturer (Pickering et al., 2005; Jani et al., 2006). The oscillometric technique offers a number of advantages regarding ambulatory blood pressure monitoring. Aside from being less susceptible to external noise, the cuff can be placed and removed easily by the patient. Moreover, this technique requires no transducer to be held manually over the brachial artery, reducing any effects resulting from incorrect placement of the cuff around the arm.

The pressure oscillations, caused by the pulsation of the blood through the arteries, are recorded using a pressure transducer connected to an inflatable cuff. Typically, the cuff is first inflated beyond the expected SBP. The cuff is then deflated at a steady rate while the transducer records the pressure oscillations, as seen in the upper waveform of Figure 2.4. The pressure

introduced by the inflated cuff pushes inwards on the exterior of the arterial wall, as opposed to the arterial pressure where the flow of blood through the arteries pushes outwards on the interior of the arterial wall. When the cuff pressure is equal to the arterial pressure, the net pressure on the arterial wall is zero and there is no deformation in the shape (Drzewiecki, Hood and Apple, 1994). The blood pressure recorded when the cuff pressure is equal to the arterial pressure is MAP. Consequently, MAP is equal to the cuff pressure recording at the point where the pressure oscillations reach a maximum amplitude (Shahriari et al., 2003; Chen et al., 2009). SBP and DBP are calculated using empirically derived "characteristic ratios" in addition to the measured MAP as described below (Geddes et al., 1982; Chen et al., 2009).

2.9.1 Maximum Amplitude Algorithm

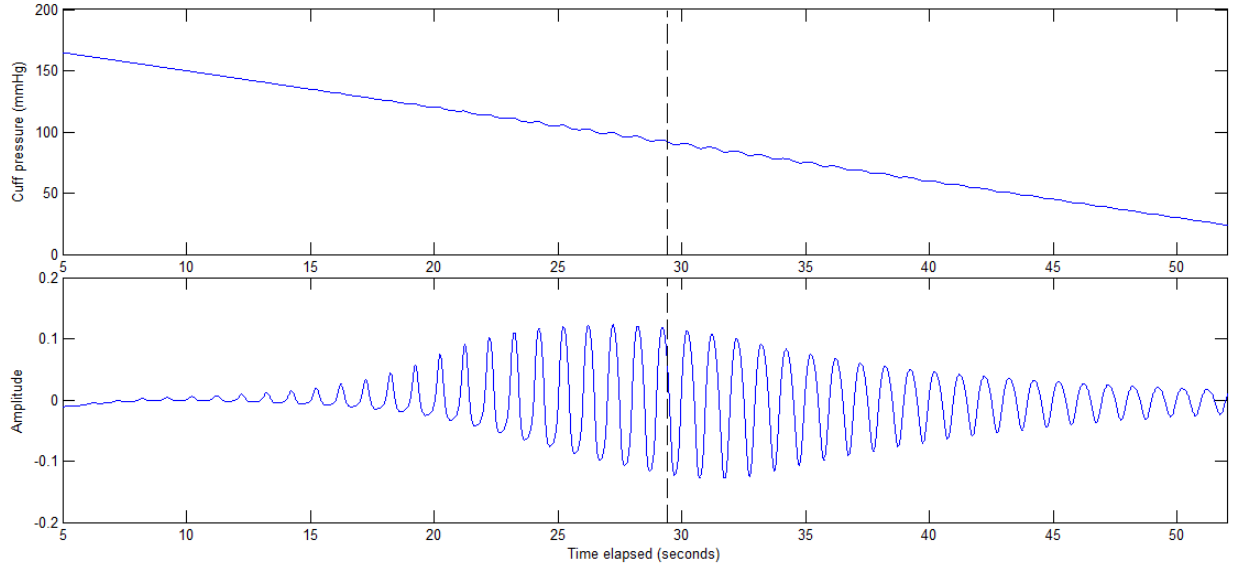


Figure 2.4 Top: Cuff pressure recorded by an oscillometric monitor during the deflation process. Bottom: Oscillometric waveform obtained by filtering the recorded cuff pressure waveform. The dashed line corresponds to the point of maximum oscillation amplitude.

The dashed line in Figure 2.4 corresponds to the point where the amplitude of the oscillations reaches its maximum, hence the name Maximum Amplitude Algorithm (MAA). This point is also where the cuff pressure is equal to MAP. The cuff pressure waveform, on top,

is the pressure signal recorded by the pressure sensor in the cuff during the deflation process. The oscillometric waveform at the bottom is obtained by filtering and detrending the cuff pressure waveform to reflect the intra-arterial blood pressure oscillations (Amoore, 2006). Detrending is performed to remove the lower frequency components in the signal that are due to the deflation procedure. Figure 2.5 shows an example of MAA. The envelope of the oscillometric waveform, as seen at the bottom of Figure 2.5, is determined by locating all the peaks and troughs on the oscillometric waveform and calculating their peak-to-peak amplitudes (Chen et al., 2009). Using MAA, the pulse where the envelope reaches its maximum amplitude is determined and mapped back to its corresponding pulse on the cuff pressure recording to determine MAP, which is equal to the cuff pressure at that point (Drzewiecki et al.1994).

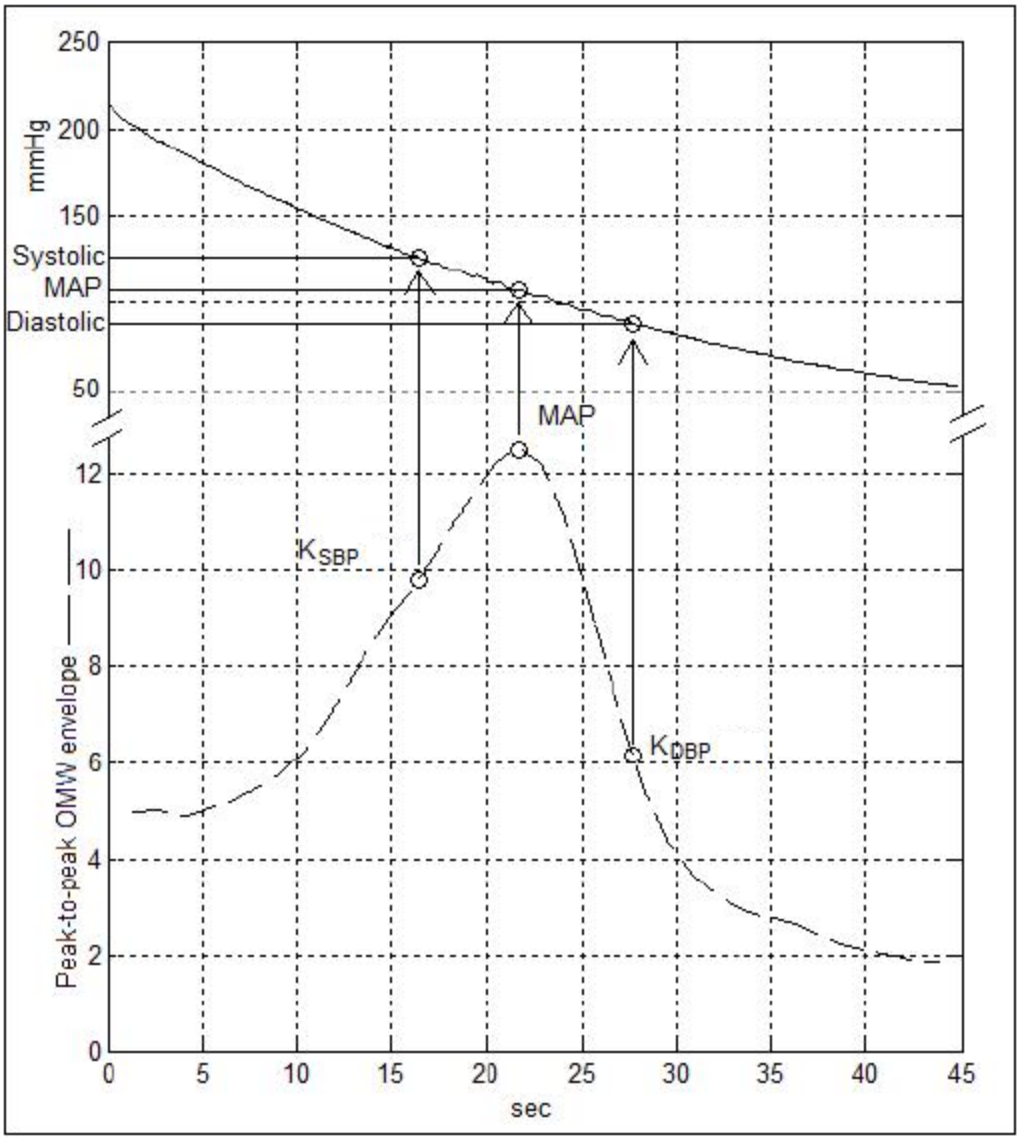


Figure 2.5 Top: Cuff pressure waveform recorded by cuff pressure sensor. Bottom: Envelope of the oscillometric waveform. The peak point on the envelope corresponds to MAP. Both K_{SBP}, corresponding to SBP, and K_{DBP}, corresponding to DBP, are determined based on an empirically-determined ratio of the envelope's peak (adapted from Chen et al., 2009 with modification).

2.9.2 Challenges

One of the challenges encountered with the use of the oscillometric technique is that the oscillations' amplitudes depend on other factors in addition to the blood pressure itself, such as the stiffness of the arteries. MAP in older patients suffering from arterial stiffness and wide pulse pressure tends to be significantly underestimated (van Montfrans, 2001). For this reason, the

monitor in use should be calibrated and validated on each patient before taking the measurement, which is rarely done in practice.

Often, a single reading is recorded before a decision regarding treatment is made. The existence of a large error could lead the physician to recommend an inappropriate course of treatment. Such errors can be reduced by taking several consecutive readings and using the mean SBP and DBP values, but this is not always done in clinical practice and rarely done in the context of self-care at home.

Oscillometric devices use proprietary algorithms, which vary from device to device, to estimate MAP, SBP and DBP. Unfortunately, many of these algorithms assume that the heartbeat rate remains stable (Bern, 2007) and therefore require several successive oscillations of similar amplitude to produce accurate results (Anastas et al., 2008). The amplitude of the oscillations fluctuates rapidly in the presence of atrial fibrillation due to the increased variability in cardiac stroke volumes (Giuliano, 2005). The increased difficulty of locating the MAP, SBP and DBP pressure points due to the rapid amplitude changes is a challenge to existing algorithms (Anastas et al., 2008).

Furthermore, Lamb and colleagues (2010) state that no oscillometric device has been validated in patients with atrial fibrillation through recognized protocols. The accuracy of oscillometric monitors is determined by comparing the difference between the readings obtained by the oscillometric monitor and the reference ones obtained through auscultation using a mercury sphygmomanometer, or less commonly, though the invasive intra-artertial method. The American National Standard protocol, ANSI/AAMI SP10, recommends an oscillometric monitor for use if the Mean of the difference Error values (ME) is within ±5 mmHg of the reference readings, with a STandard Deviation of the Error (STDE) of less than 8 mmHg (Association for

the Advancement of Medical Instrumentation, 2003). The British Hypertension Society (BHS) recommends an oscillometric monitor for use if it achieves a BHS grade of A or B (Table 2.1) for both SBP and DBP readings (British Hypertension Society, 2012).

As shown in Table 2.2, the BHS grade is based on the percentage of the readings obtained that are within an absolute difference error range relative to the reference method's readings. For an oscillometric monitor to achieve grade A, at least 60%, 85% and 95% of its SBP and DBP readings have to be within 5 mmHg, 10 mmHg or 15 mmHg of the reference readings, respectively.

	Absolute Difference Error (mmHg)		
	≤ 5	≤ 10	≤ 15
Grade	**Percentage of readings required (%)**		
A	60	85	95
B	50	75	90
C	40	65	85
D	Lower than Grade C		

Table 2.2 BHS Grading Criteria (adapted from Reinders et al., 2005).

2.10 Prior Art

For the measurement of blood pressure in the presence of atrial fibrillation, Stewart and colleagues compared the accuracy of two electronic sphygmomanometers, Takeda UA-751 and Copal UA-251, and two ambulatory blood pressure monitors, Accutracker 1 and SpaceLabs 90207, and used a mercury sphygmomanometer as a reference (Stewart et al., 1995). The devices were compared using the same arm, by sequentially switching with the mercury sphygmomanometer which was treated as the gold standard. The performance of all four devices was evaluated in one sitting in all thirty patients with atrial fibrillation. The Copal UA-251, a

device that uses a microphone to detect Korotkoff sounds, was found to be within 5 mmHg of the mercury sphygmomanometer readings in 70% of readings. The device passed the validation and recommendation criteria, but unfortunately, production of this particular monitor has been discontinued.

A study by Shahriari et al. (2003) was conducted to evaluate the performance of five oscillometric blood pressure monitors: three of which use wrist cuffs and two use upper arm cuffs. Every monitor's blood pressure measurement procedure was preceded by an auscultatory reading and followed by another, for each of the seventy-two patients recruited. Table 2.3 shows a summary of the differences in the SBP and DBP readings between each monitor and the reference readings obtained manually through auscultation, expressed in terms of ME ± STDE, and whether each monitor satisfies the ANSI/AAMI protocol requirements (Association for the Advancement of Medical Instrumentation, 2003). Even though three of the monitors passed the AAMI criteria for DBP readings, none was able to pass for SBP readings.

Monitor	Cuff placement	ME ± STDE (mmHg)		AAMI Approval
		SBP	DBP	SBP / DBP
Omron R4	Wrist	1.2 ± 12.6	-0.5 ± 8.0	No / Yes
A&D UB 322	Wrist	-8.5 ± 15.1	-4.4 ± 8.2	No / No
Braun	Wrist	-3.6 ± 18.0	-8.3 ± 6.9	No / Yes
A&D UA 777	Upper Arm	-5.5 ± 8.3	-6.8 ± 6.8	No / No
Omron M4	Upper Arm	-5.5 ± 8.5	-3.0 ± 5.9	No / Yes

Table 2.3 Summary of differences between the readings of five different oscillometric blood pressure monitors and readings obtained through auscultation, expressed as the Mean of the Errors (ME) ± STandard Deviation of the Errors (STDE), and whether each device meets the ANSI/AAMI protocol requirements for AAMI approval or not (adapted from Shahriari et al., 2003).

None of the oscillometric blood pressure monitors investigated in this study managed to score a passing grade of A or B according to the BHS protocols either. The right-most column in Table 2.4 shows what BHS grade each device scored for SBP and DBP.

Absolute Difference Error (mmHg)				
	≤ 5	≤ 10	≤ 15	
Monitor	**Percentage of SBP / DBP readings**			**SBP / DBP BHS Grade**
Omron R4	42 / 51	68 / 86	82 / 94	C / B
A&D UB 322	23 / 40	49 / 75	65 / 90	D / C
Braun	38 / 31	51 / 71	69 / 90	D / D
A&D UA 777	39 / 36	67 / 70	89 / 90	D / D
Omron M4	36 / 60	69 / 89	86 / 99	D / A

Table 2.4 Percentages of SBP and DBP readings with absolute difference errors of 5, 10 and 15 mmHg from the reference readings for each monitor and the BHS grade achieved (adapted from Shahriari et al., 2003).

Based on these results, Shahriari et al. (2003) recommend the use of mercury sphygmomanometers because no existing oscillometric blood pressure monitor can provide accurate blood pressure measurements in atrial fibrillation patients.

Jani and colleagues (2006) compared the accuracy of Omron HEM-750 CP readings with readings from a mercury sphygmomanometer, in patients with atrial fibrillation. This device is an oscillometric device that is able to detect changes in pressure oscillations that occur as a result of the arterial wall movement beneath the inflated cuff. All readings obtained were within 15 mmHg from the mercury sphygmomanometer readings so the authors claimed that it can be used in a clinical setting in patients with atrial fibrillation, along with multiple readings with a mercury sphygmomanometer.

The differences between the readings obtained using automated oscillometric monitors and the ones obtained through manual auscultation in patients hospitalized for critical care, according to Bern and colleagues (2007), is because automated monitors detect the vibrations, or oscillations, of the arterial wall rather than the sounds generated by the arterial blood flow. A study of 126 inpatients was conducted to compare the blood pressure readings of a Dinamap VSM 8100 automated blood pressure monitor to a manual Welch Allyn sphygmomanometer. SBP and DBP readings from each device were obtained sequentially for every patient, with a time gap between each measurement lasting from two to five minutes. Analysis of the results shows that the differences in SBP readings obtained using both methods are statistically significant, and while the differences in DBP readings are not, there is a wide range of DBP differences. Hence, the authors recommend using the auscultation technique more frequently where the blood pressure readings may affect medication-based decisions (Bern et al., 2007).

Another study (Anastas et al., 2008) was designed to assess the performance of an automated oscillometric blood pressure monitor, Welch Allyn Vital Signs Monitor, against a manually-operated aneroid sphygmomanometer, Welch Allyn Model 7670-01. The sample recruited for the study was limited to fifty-three patients hospitalized with atrial fibrillation, which, as discussed above, diminishes the accuracy of automated blood pressure measurements (van Montfrans, 2001). The two blood pressure measurement procedures were performed sequentially with a time gap of sixty seconds between each measurement, starting with a randomly chosen procedure and then alternating. The authors reported significant differences between the automated monitor's readings and the ones manually-obtained through auscultation (Table 2.5).

Absolute Difference Error	Percentage of readings		BHS Grade SBP / DBP
	SBP (%)	DBP (%)	
≤ 10 mmHg	51	85	D / A
≤ 15 mmHg	85	92	C / B

Table 2.5 Summary of the performance of the Welch Allyn monitor, showing the percentage of SBP and DBP readings with absolute difference errors of 10 mmHg and 15 mmHg from the reference readings and the BHS grade achieved (adapted from Anastas et al., 2008).

The authors conclude that automated blood pressure monitors cannot estimate DBP and heart rate with reliable accuracy in the presence of atrial fibrillation (Anastas et al., 2008).

Lamb and colleagues compared two commonly used oscillometric monitors, Omron HEM 7111 AC and Welch-Allyn 52000 series oximeter, in the presence of atrial fibrillation (Lamb et al., 2010). The Omron device is a common home blood pressure monitor while the Welch-Allyn oximeter can be found in hospitals. One hundred and five subjects were recruited for the study, fifty one of which had atrial fibrillation while the rest served as controls. Blood pressure readings were taken simultaneously using an oscillometric device on one arm and a mercury sphygmomanometer on the other, for each monitor. In the subjects with atrial fibrillation, a quarter of the readings were found to be different from the mercury sphygmomanometer readings by more than 10 mmHg, which fails to meet the American National Standard and the British Hypertension Society validation requirements.

Chapter 3. Methodology

In this chapter, the equipment used in this thesis are briefly described, along with why each one was chosen. Both datasets used (healthy subjects and cardiovascular patients) are described as well, in terms of the subject characteristics and the data acquisition procedure. Finally, and most significantly, the proposed approach is described in detail, along with figures, to explain every step of the algorithm.

3.1 Equipment used to Estimate Blood Pressure

3.1.1 BpTRU BPM-200

The BpTRU BPM-200 (BpTRU Medical Devices, Coquitlam, BC) is an automated blood pressure monitor that estimates Systolic Blood Pressure (SBP) and Diastolic Blood Pressure (DBP) noninvasively. The cuff, wrapped around the upper arm, inflates then deflates automatically at a steady rate of 4 mmHg/sec. Since this device probably uses the oscillometric technique, during deflation Mean Arterial Pressure (MAP) is understood to be measured directly from the cuff pressure waveform recorded by the transducer embedded in the cuff. SBP and DBP are then calculated using a proprietary algorithm that uses empirically-derived ratios on MAP pulse amplitudes (Mattu et al., 2004). This monitor is capable of taking multiple recordings automatically over six blood pressure measurements, where the first recording is discarded and the average of the five remaining SBP and DBP readings is displayed. It has been proposed that discarding the first recording and displaying the average SBP and DBP may reduce the undesirable impact of the white coat effect (Graves et al., 2003).

3.1.2 Omron HEM790IT

Another oscillometric monitor utilized in this study is the Omron HEM790IT (Omron Healthcare Inc., Bannockburn, IL), a non-invasive automated blood pressure monitor validated

according to the British Hypertension Society (BHS) protocols for home and office use that is capable of detecting the presence of arrhythmias (Coleman et al., 2008). In addition to using oscillometry, this monitor is equipped with TruRead mode, where the average SBP and DBP of up to three consecutive readings can be calculated and displayed automatically (Omron Healthcare, n.d.).

3.1.3 InBeam Prototype

This non-invasive oscillometric blood pressure monitor was developed at the School of Electrical Engineering and Computer Science (EECS) of the University of Ottawa (Ahmad et al., 2010). The main components of the monitor are the pressure transducer (Vernier Pressure Transducer BPS-BTA, Beaverton, OR), the analog ElectroCardioGram (ECG) amplifier (Texas Instruments, Dallas, TX), a miniature direct-current air pump, and a means of controlling the pressure release valve manually. The analog voltage outputs produced by the pressure transducer and the ECG amplifier are conditioned, buffered, and then sampled by an analog-to-digital converter before being transmitted to a personal computer via a Universal Serial Bus (USB) cable.

The cuff is wrapped around the upper portion of the subject's left arm and a wristband is worn around the wrist of the other arm. A dry flexible electrode made from conductive fabric is embedded in the inner side of the cuff and the dry electrode embedded in the wristband forms a second electrode for detecting the ECG signal. Following the inflation of the cuff, cuff pressure and ECG signals are collected simultaneously during deflation.

3.1.4 InBeam V2

Ahmad and colleagues developed a novel approach for blood pressure measurement that makes use of the single-lead ECG signal. The cuff inflates to a certain point then begins deflating slowly at a constant rate of 3 mmHg/sec as the ECG and cuff pressure signals are being recorded. Off-line signal processing is performed to obtain a filtered version of the cuff pressure signal that reflects the arterial blood pressure oscillations throughout the deflation process. Rather than using the Maximum Amplitude Algorithm (MAA) to calculate MAP, SBP and DBP, this method makes use of the Pulse-Transit Time (PTT) of each pulse. The PTT, also referred to as Pulse-wave Propagation Time, is the time it takes for a pressure pulse produced by the heart to propagate from the aorta to a certain point in the arteries, typically where the cuff sensor is placed. This can be measured by finding the time difference between the heart's electric impulse, represented by the R-peak of the ECG waveform, and a point on the pulse waveform, such as the trough or the crest. Recordings from ten subjects were analyzed in a pilot study that showed an improved accuracy in the estimation of MAP, SBP and DBP when compared to an MAA-based FDA-approved blood pressure monitor (Ahmad et al., 2012).

The non-invasive oscillometric monitor, InBeam V2, a newer version of the InBeam prototype, records cuff pressure oscillations and ECG signals in a similar fashion to the InBeam prototype; However, unlike its predecessor, it employs a different flexible cuff in addition to a rigid, rather than flexible, ECG electrode (Ahmad et al., 2012).

3.2 Dataset from Healthy Subjects

3.2.1 Participants

A set of data was obtained at the University of Ottawa, by a research associate, Dr. Saif Ahmad (Ahmad, 2012), from ten healthy subjects with ages ranging from 24 to 63 years, of whom six are male and four are female. The subjects had no history of cardiovascular or respiratory diseases. This study was authorized by the University of Ottawa Research Ethics Board and all participants provided written informed consent (Ahmad et al., 2010).

3.2.2 Experimental Procedure

During measurement, the subjects were seated in an upright position that maintained the arms at the same horizontal level as the heart. The subjects were asked to breathe normally and refrain from crossing their legs or making any unnecessary movements. Each of the ten subjects was asked to attend three sessions over three different days. Five measurements, per monitor, were recorded in a single day for every subject, producing a total of 300 recordings; 150 measurements from the InBeam monitor paired with another 150 measurements from the Omron monitor.

The Omron cuff was wrapped around the subject's upper right arm and the InBeam cuff around the opposite upper arm. The Omron monitor recorded the first measurement and this was then immediately followed by the InBeam monitor, producing a pair of measurements. Three minutes later, the Omron monitor would take a second measurement, again immediately followed by its counterpart - the InBeam monitor. This process was repeated, with three-minute intervals as recommended by AAMI (Association for the Advancement of Medical Instrumentation, 2003), until five pairs of measurements were recorded; five from Omron paired with five from InBeam, adding up to ten measurements from every subject on a single day. Each

subject would later come in on a second and third day to go through the same procedure. The thirty measurements (fifteen from Omron plus fifteen from InBeam) collected from each of the ten healthy subjects, over three different days, make up the first dataset of 300 measurements (150 from Omron plus 150 from InBeam) for this study.

The Omron and the InBeam monitors operate by inflating their respective cuffs to a point beyond an expected SBP before deflating at a steady rate. Both monitors are embedded with transducers that record the amplitude of the oscillations during deflation and record an oscillometric cuff pressure waveform, although we do not have access to the waveform recorded by the Omron monitor.

Synchronized ECG data was obtained by the InBeam monitor, in addition to the oscillometric cuff pressure recordings, for the purposes of a different study. The ECG data was recorded using dry flexible electrodes, made of conductive fabric, embedded in the InBeam cuff and in the wristband. For the purposes of this study, from InBeam, only the cuff pressure recordings were used.

The procedure undertaken to record data from each subject, per session, can be summarized in the following steps:

1) Place the Omron cuff around right arm, the InBeam cuff around left arm and the InBeam wristband around the left wrist.

2) Ask subject to breath normally and refrain from crossing their legs or making any unnecessary movements.

3) Take Omron recording.

4) Take InBeam recording as soon as Omron recording is complete.

5) Wait three minutes.

6) Repeat steps 3 and 4 four times.

3.3 Dataset from Patients

3.3.1 Participants

In coordination with a cardiac specialist at the Ottawa Cardiovascular Center, Dr. Sanjeev Chander, M.D., cuff pressure and ECG waveforms were collected by a research associate, Dr. Saif Ahmad (Ahmad, 2012). Thirteen patients, eight female and five male, with a mean age of about 73 years and ages ranging between 46 and 86 years participated in this study, with one additional patient excluded from the study. This study was authorized by the University of Ottawa Research Ethics Board and in addition to signing an informed consent form (APPENDIX A), all patients answered a participant questionnaire (APPENDIX B) disclosing gender, age, height, weight and history of cardiovascular diseases.

As seen in Table 3.1, the answers to the questionnaire revealed that ten out of the thirteen patients suffered from some form of atrial fibrillation, seven were hypertensive and out of those, four suffered from both atrial fibrillation and hypertension.

Patient ID #	Sex	Age (years)	Height (m)	Weight (Kg)	Cardiovascular Condition
1	F	80	1.60	81	Controlled atrial fibrillation
2	F	65	1.57	47	Controlled atrial fibrillation
3	F	46	1.63	109	Controlled hypertension
4	F	72	1.62	105	Controlled hypertension
5	F	71	1.63	77	Controlled hypertension
6	F	N/A	N/A	N/A	Patient was excluded from study
7	F	76	1.64	72	Controlled atrial fibrillation
8	M	70	1.83	86	Controlled atrial fibrillation Controlled hypertension
9	M	79	1.78	86	Controlled atrial fibrillation Hypotension
10	F	80	1.60	63	Mild atrial fibrillation
11	M	77	1.75	93	Controlled atrial fibrillation Controlled hypertension
12	F	72	1.63	88	Controlled atrial fibrillation
13	M	85	1.87	88	Controlled atrial fibrillation Controlled hypertension
14	M	72	1.73	109	Controlled atrial fibrillation Controlled hypertension

Table 3.1 Summary of the responses to the participant questionnaire (APPENDIX B).

It should be noted that in the dataset of patients, 27 recordings exhibited episodes of atrial fibrillation during the blood pressure measurement procedure.

3.3.2 Experimental Procedure

During the measurement, the patients were seated comfortably in a chair, while maintaining the arms at the same level as the heart and were asked to remain calm and breathe normally with minimum movement and without crossing their legs. Each measurement was completed in thirty to seventy seconds and the entire session, for each patient, was completed in a single day and took about one hour.

The procedure consisted of two phases. The first phase was performed with the BpTRU cuff wrapped around the upper right arm and the InBeam V2 cuff wrapped around the upper left arm. The BpTRU monitor recorded the first measurement then was followed immediately by the InBeam V2 monitor, producing a pair of measurements. In the same fashion, another pair of measurements was recorded after waiting for the recommended period of three minutes (Association for the Advancement of Medical Instrumentation, 2003). Three minutes later, one more pair of measurements was recorded. This phase produced six measurements (three from BpTRU paired with three from InBeam V2) that were recorded with three-minute intervals between each pair of measurements. For the second phase of the procedure, both the BpTRU and InBeam V2 cuffs were removed and placed around the opposite arms; BpTRU around the upper left arm and InBeam V2 around the upper right arm. By following the same procedure as the first phase, six more measurements (three from BpTRU paired with three from InBeam V2) were recorded with three-minute intervals, adding up to a total of twelve measurements (six from BpTRU and six from InBeam V2) for each patient.

The twelve measurements (six from BpTRU plus six from InBeam V2) collected from each of the thirteen patients make up the second dataset of 156 measurements (78 from BpTRU plus 78 from InBeam V2) for this study.

Both monitors operated by inflating the cuff beyond an expected SBP and then deflating to below an expected DBP at a steady rate for every recording. The transducer embedded in each monitor records the amplitude of the oscillations produced by the deflation of the cuff and record an oscillometric cuff pressure waveform, although we do not have access to the waveform recorded by the BpTRU monitor.

Synchronized ECG data was obtained by the InBeam V2 monitor, in addition to the oscillometric cuff pressure recordings, for the purposes of a different study. The ECG data was recorded using dry flexible electrodes embedded in the InBeam V2 cuff and by holding the dry electrode on the device box. For the purposes of this study, from InBeam V2, only the cuff pressure recordings were used.

The procedure undertaken to record data from each patient can be summarized in the following steps:

1) Place the BpTRU cuff around right arm and the InBeam V2 cuff around left arm.
2) Ask patient to hold the rigid electrode on the InBeam V2 device box.
3) Ask subject to remain calm, breath normally and refrain from unnecessary movement.
4) Take BpTRU recording.
5) Take InBeam V2 recording as soon as BpTRU recording is complete.
6) Wait three minutes.
7) Repeat steps 4 - 6 twice.
8) Remove both cuffs and replace them around the opposite arms.
9) Take BpTRU recording.
10) Take InBeam V2 recording as soon as BpTRU recording is complete.
11) Wait three minutes.
12) Repeat steps 9 - 11 twice.

3.4 Ratio-Based Analysis

This thesis proposes the use of a ratio-based method for calculating blood pressure parameters non-invasively using oscillometric devices, as opposed to the common and widely used amplitude-based MAA method. Upon surveying the literature available at the time, no mention of the use of any ratio-based algorithm for calculating MAP in people affected by cardiac arrhythmias was found. Instead, many research papers and studies (Sykes D. et al., 1990; Stewart M. et al., 1995; Kaliujnaya V. and Kalyuzhny S., 2005; Jani B. et al., 2006; Anastas Z. et al., 2008; Lamb T. et al., 2010; Stergiou G.S. et al., 2012) only criticized the widespread use of MAA in determining MAP, SBP and DBP in people afflicted with atrial fibrillation, but provided no alternative other than recording several measurements and using the average values. None-the-less, it appears that in both healthy humans and atrial fibrillation patients the amplitude, from a baseline, of the peak of the pulse where the cuff pressure is equal to MAP is approximately twice that of the amplitude of the trough, from the same baseline.

MAP is a time-weighted average of blood pressure, rather than a basic arithmetic average of SBP and DBP, because it spends about twice as long in diastole than systole, at normal resting heart rates (Pocock and Richards, 1999). This is why MAP has long been estimated using Equation 2.3 (Geddes et al., 1982). Equation 2.3 points out that MAP is a value that falls between SBP and DBP, and is twice as far from SBP than DBP. For example, in the case of the typical 120/80 mmHg, where SBP is 120 mmHg and DBP is 80 mmHg, Equation 2.3 would produce MAP = 93.33 mmHg. The difference between SBP and MAP would be 26.67 mmHg, that is twice the difference between MAP and DBP which is 13.33 mmHg. Based on this, the ratio value of 2 was selected.

3.4.1 Proposed Algorithm

Once the algorithm opens the file produced by the monitoring device and reads its contents, the time stamp and the cuff pressure signal are stored as arrays of values. The cuff pressure waveform (Figure 3.1), recorded during deflation, is time-interpolated then passed through a 2nd order Butterworth band-pass filter with cut-off frequencies of 0.5 Hz and 25 Hz to remove the trend line and suppress higher frequency noise from the cuff pressure waveform. This operation produces an oscillometric waveform like the one in Figure 3.2.

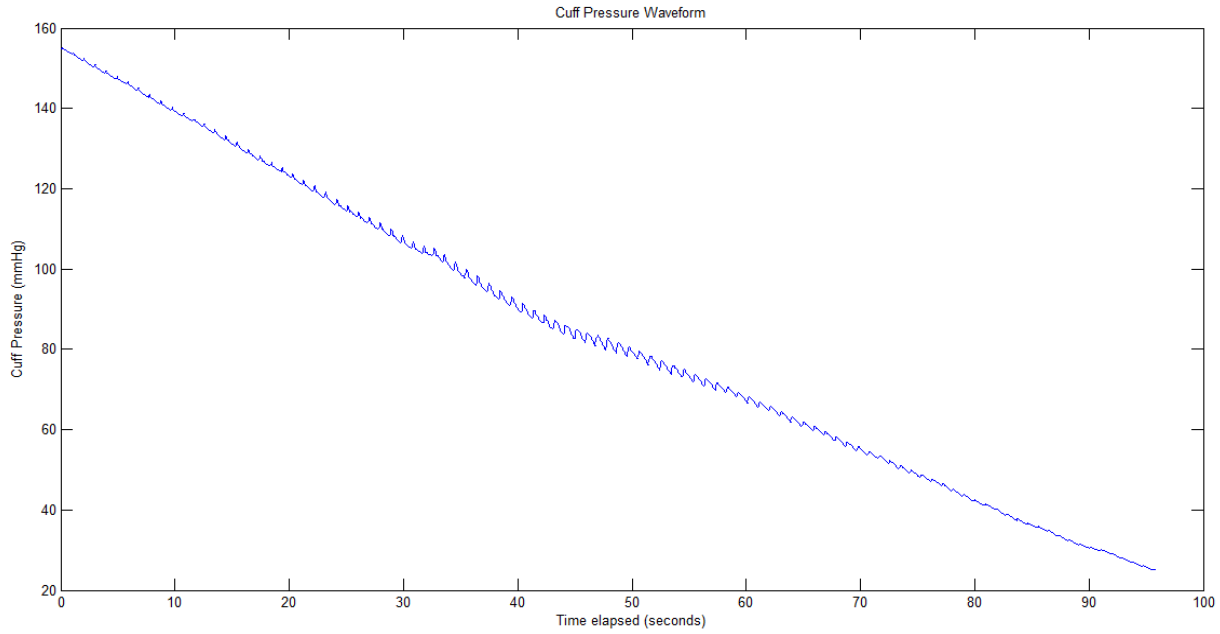


Figure 3.1 Cuff pressure waveform, recorded by the InBeam monitor from one of the healthy subjects, showing the cuff pressure as a function of time.

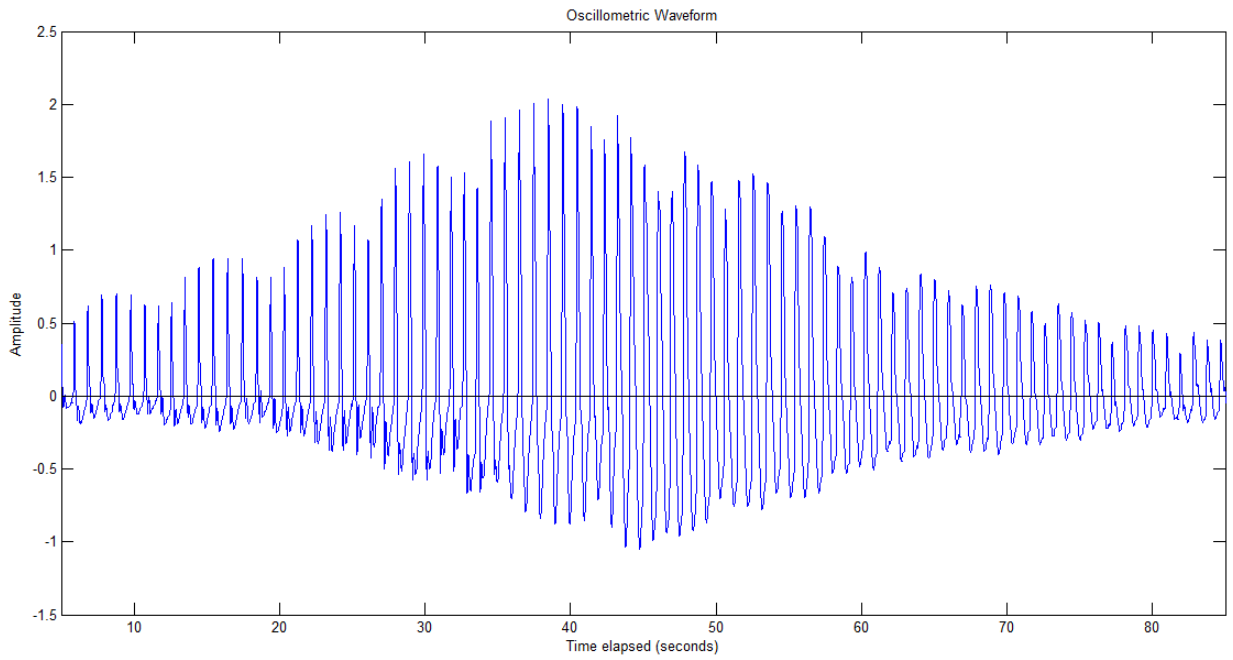


Figure 3.2 Oscillometric waveform obtained, as a function of time, by passing the cuff pressure waveform from Figure 3.1 through the band-pass filter described in the text.

Next, a peak detection algorithm is used to find and locate the peaks and troughs on the oscillometric waveform. A modified version of an open-source peak detection algorithm, written in Matlab (The MathWorks Inc., Natick, MA) and released to the public domain (Billauer, 2012), is used to find the local maxima and minima on the oscillometric waveform as well as their indices, effectively determining the location of every pulse's peak and trough and the amplitude of each peak and trough from the baseline of the oscillometric waveform that is obtained after detrending the cuff pressure waveform by passing it through the band-pass filter. A peak is typically identified as the point with the highest value that is preceded and followed by points of lower values. To avoid identifying the dichrotic notch, that in itself has its own peak and trough, as a peak, a threshold value (Δ = 20% of the highest point on OMW) is needed. Now, only the peaks preceded by a value that is lower by at least Δ are detected. Figure 3.3 shows the effects of using a threshold value. Without a threshold (Figure 3.3, top), every

dichrotic notch would be detected as a peak and a trough and as a result, would be treated as an individual pulse.

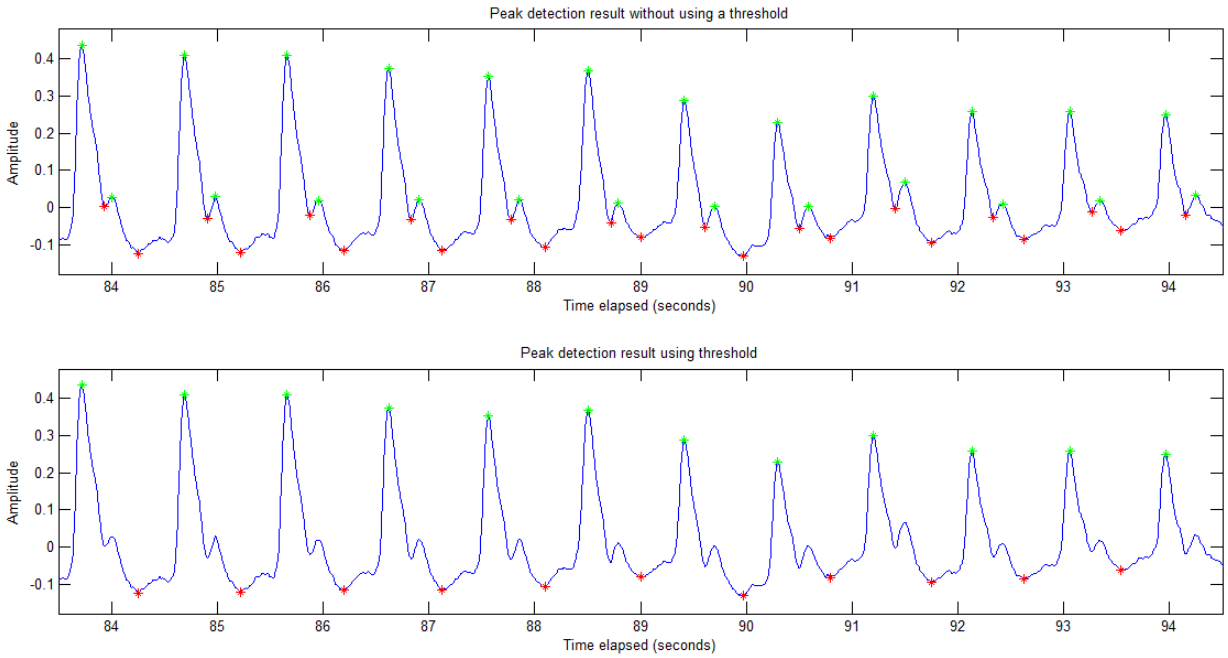


Figure 3.3 Effect of using a threshold value in the peak detection algorithm. Green markings indicate peaks and red markings indicate troughs. Top: Result of the peak detection algorithm without using a threshold. Bottom: Result of the peak detection algorithm using a threshold.

The peaks and troughs of the premature pulses that occur in atrial fibrillation were discarded. In order to do that, the 3-point moving average of the peaks' amplitudes was calculated using Matlab's *tsmovavg* function, in addition to the STandard Deviation (STD) of the peaks' amplitudes. All peaks that had amplitudes lower than the moving average by more than 0.2*STD were considered as premature beats and were discarded. The value of the factor (0.2) was determined by visually examining its effect on the signal. Figure 3.4 shows the OMW of a patient with atrial fibrillation before and after the cleaning procedure that discards the peaks of the premature pulses.

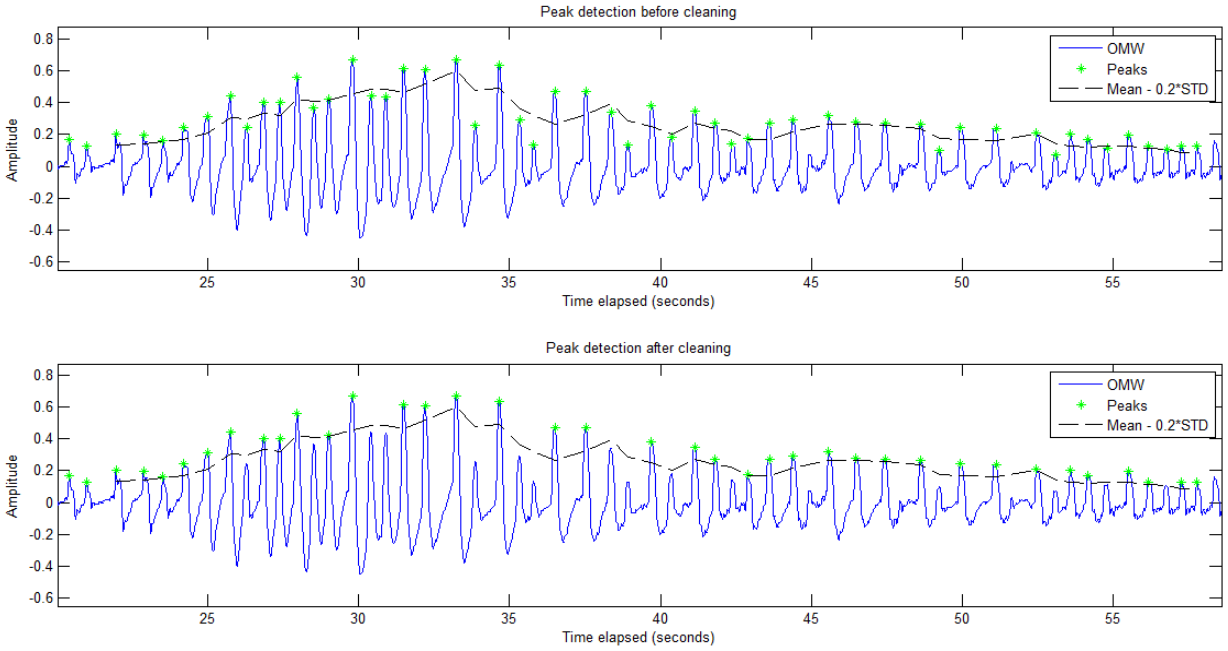


Figure 3.4 Peak detection on OMW of an atrial fibrillation patient. The green markings represent the peaks identified and the dashed black line represents the 3-point moving average minus 0.2 times the STD of the peaks' amplitudes. Top: All the peaks detected before cleaning. Bottom: Peaks with amplitude lower than the moving average by more than 0.2*STD were removed.

Figure 3.5 below shows the result of applying the peak detection algorithm to the OMW from Figure 3.2. In this case, the subject had a regular heart rhythm.

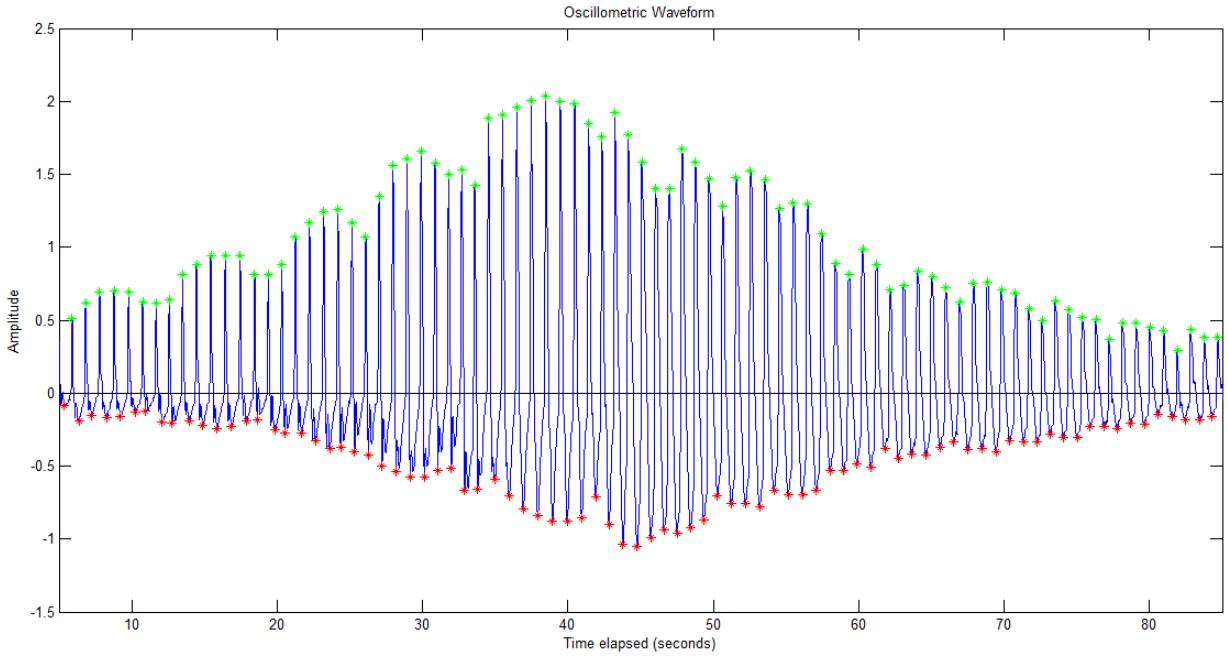


Figure 3.5 Oscillometric waveform obtained after applying the peak detection algorithm to the oscillometric waveform from Figure 3.2. The green markings indicate the peaks, the red markings indicate the troughs and the black line indicates the baseline.

The ratio of a pulse is calculated by dividing the pulse's amplitude of its peak (highest value) from the baseline (the zero level of the oscillometric waveform) by the amplitude of its trough (lowest value) from the same baseline. In other words, for every individual pulse:

$$\textbf{Ratio} = \frac{\textbf{Amplitude of peak} - \textbf{Baseline}}{\textbf{Baseline} - \textbf{Amplitude of trough}} = \frac{|\textbf{Amplitude of peak}|}{|\textbf{Amplitude of trough}|} \qquad \text{Equation 3.1}$$

$$\textbf{Index} = \frac{I_p + I_t}{2} \qquad \text{Equation 3.2}$$

Equation 3.2 shows how the index where ratio = 2 is determined, where I_p is the index of the peak of the pulse and I_t is the index of the trough of the same pulse. Since the difference between the time when a peak is reached and the time when a trough is reached in a pulse is miniscule, the index used to extract the cuff pressure value that is determined to be equal to MAP is calculated as the mid-point between the time when the peak is reached and the time when the trough is reached.

A ratio for every pulse is calculated in the same manner, producing an array of ratio values (Figure 3.6).

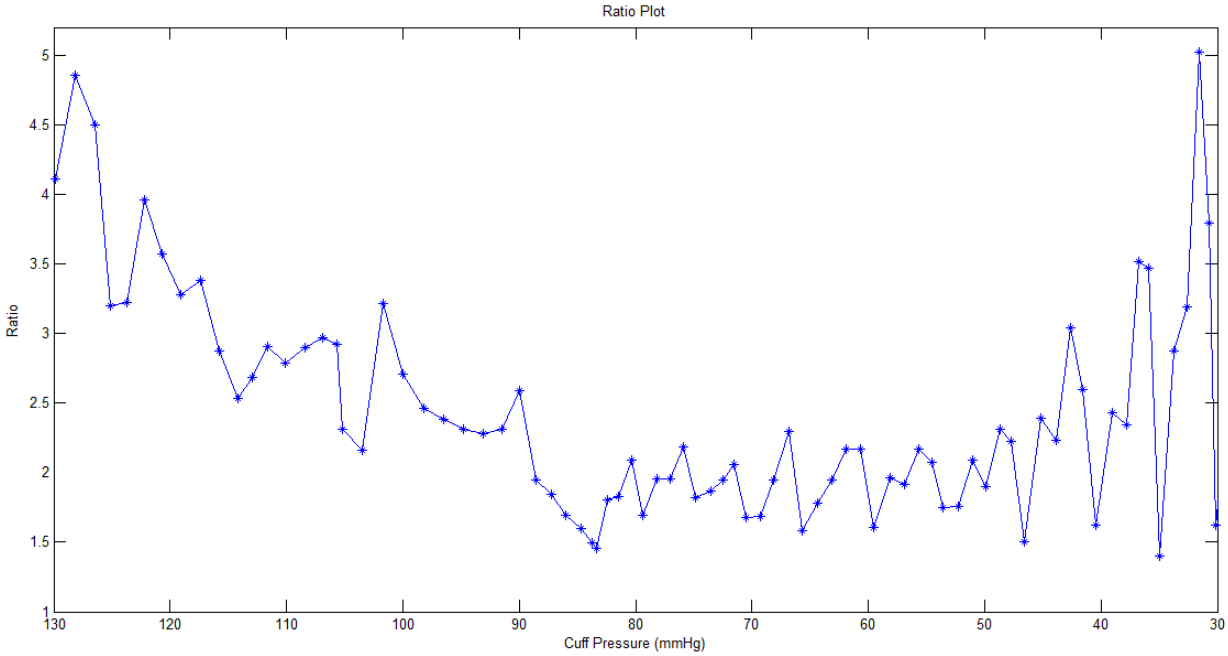


Figure 3.6 Plot of the ratio values calculated as a function of the decreasing cuff pressure.

The linearly time-interpolated array of calculated ratios is subtracted by 2, which corresponds to the ratio that is hypothesized to correspond to MAP. The zero-crossings are then located by identifying the points where the sign of the ratio values changes from positive to negative or from negative to positive. The index of the ratio value where the first zero-crossing (i.e., sign change) occurred is determined using Matlab's *find* function. The cuff pressure at that index is determined to be MAP, as seen in Figure 3.7.

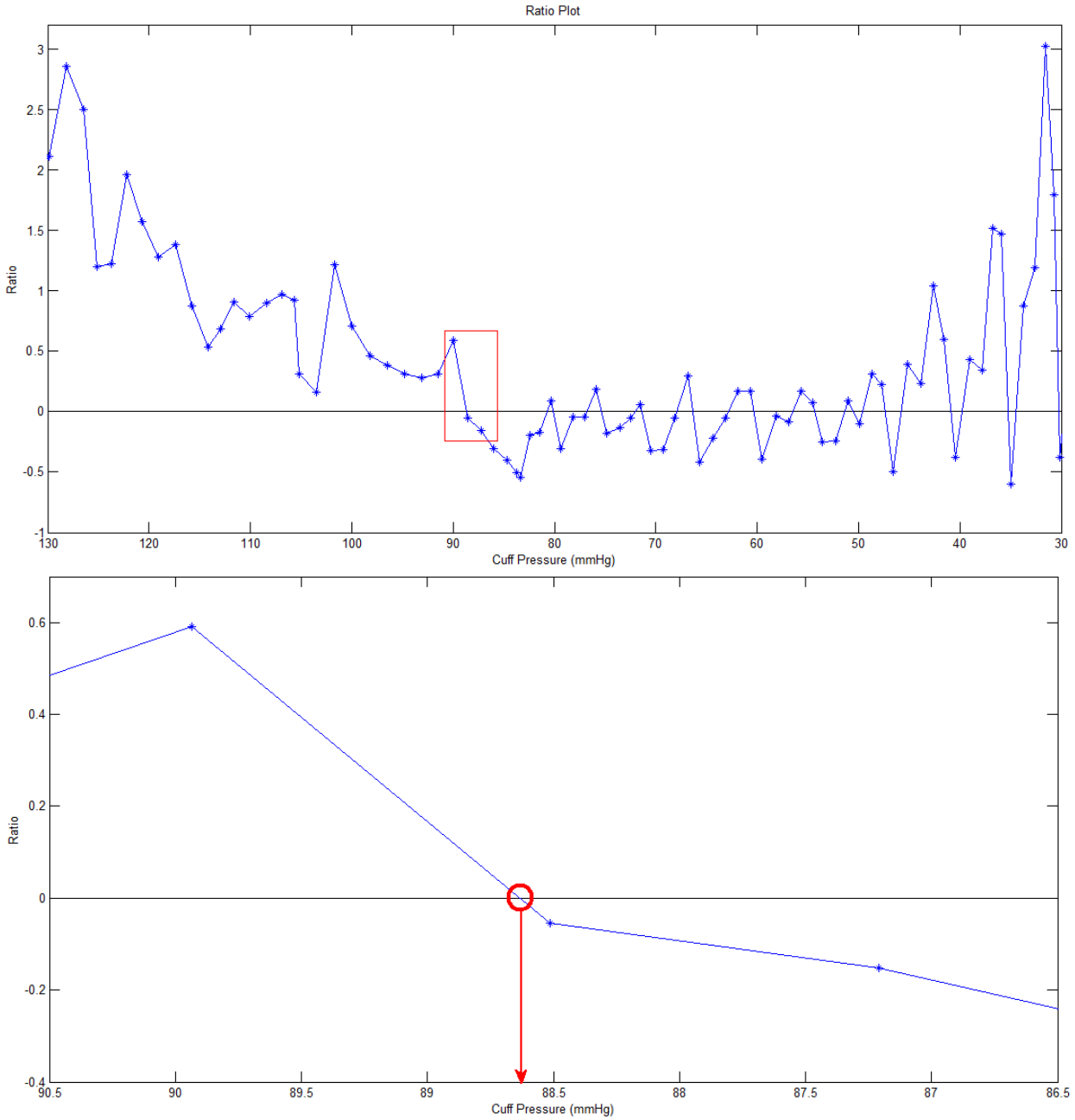


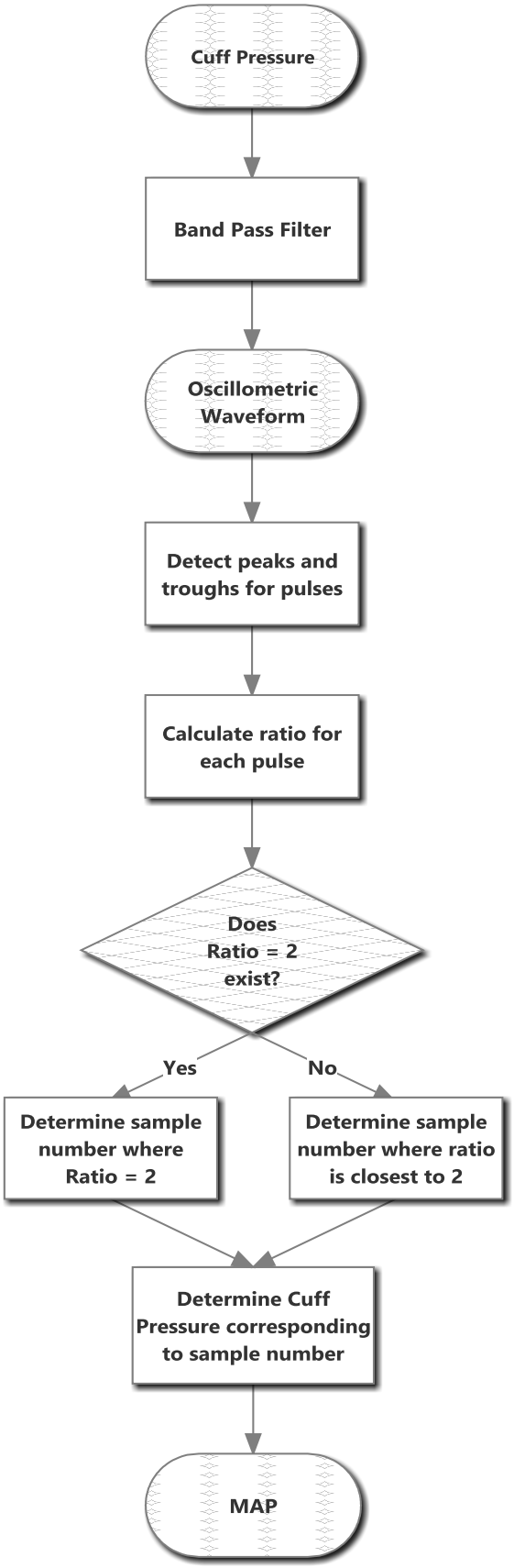
Figure 3.7 Top: Plot of the ratio values calculated as a function of the decreasing cuff pressure. Bottom: Plot obtained by zooming in on the red box in the top plot. The red circle indicates the first intersection between the ratio plot and the zero line. The red arrow is pointing to the corresponding cuff pressure.

Out of the entire set of 228 recordings (150 from the healthy dataset in addition to 78 from the patient dataset) that were processed, five ratio plots with no zero-crossings were found. This means that the ratio never reached a value of 2, either by being consistently above two or consistently below two. Those five plots were recorded from two patients, both with atrial

fibrillation. In order to fully automate the process, an *if* statement was introduced that stated that if no zero-crossing is found, the index of the ratio point closest to two is determined. This is done by, first, subtracting two from the linearly time-interpolated array of calculated ratios, then locating the lowest value (global minimum) of the absolute values. The cuff pressure at that index is determined to be MAP.

The overall procedure for determining MAP from the recorded cuff pressure waveform is summarized in Figure 3.8 below.

The proposed algorithm used to process and analyze the acquired signals was written entirely in Matlab R2011b.



First, the recorded cuff pressure values are stored as an array called Cuff Pressure.

Cuff Pressure is passed through the band-pass filter from Figure 3.2 to remove the trend line and any noise recorded during the measurement process.

The result of the filtering process is called Oscillometric Waveform.

Peaks and troughs of Oscillometric Waveform, as well as their indices, are found.

A ratio value for each detected pulse is calculated by dividing the amplitude of the pulse's peak, from the baseline, by the amplitude of the same pulse's trough, from the same baseline, creating an array.

The array of ratio values calculated is time-interpolated then subtracted by 2 and the zero-crossings are located.

YES	**NO**
If zero-crossings exist, the index of the first one is determined as the sample number where ratio = 2.	If no zero-crossings are found, the index of the ratio value closest to zero is determined as sample number where ratio is closest to 2.

The Cuff Pressure value at the index where the index is equal to the sample number is determined as MAP.

Figure 3.8 Left: Flow chart of the proposed algorithm. Right: Brief explanation of the different levels.

Chapter 4. Results and Discussion

To determine the accuracy of the proposed algorithm, the difference between Mean Arterial Pressure (MAP) determined using the proposed algorithm is compared against MAP calculated from the Systolic Blood Pressure (SBP) and Diastolic Blood Pressure (DBP) values determined by the reference devices. Since the accuracy of Equation 2.4 has been challenged repeatedly (Meaney et al., 2000; Razminia et al., 2004; Bos et al., 2007; Graf et al., 2010), three formulas were used to calculate MAP for each reference device - Equations 2.4, 2.5 and 2.6.

- $MAP_{33\%}$ is the MAP calculated using Equation 2.4 from SBP and DBP produced by the reference monitor.
- $MAP_{40\%}$ is the MAP calculated using Equation 2.5 from the same SBP and DBP produced by the reference monitor.
- MAP_{HR} is the MAP calculated using Equation 2.6 from the same SBP and DBP produced by the reference monitor.
- MAP_{ratio} is the MAP estimated using the proposed ratio-based algorithm.

The error is calculated by subtracting MAP_{ratio} from each of the three calculated MAP values. Mean Error (ME) is the average of all the reading differences calculated in a dataset. STandard Deviation of Errors (STDE) is calculated using the reading differences of an entire dataset. ME and STDE are used by the Association for the Advancement of Medical Instrumentation (AAMI) ANSI/AAMI SP-10 protocol for accuracy validation (Alpert et al., 2010). Mean of Absolute Errors (MAE) is the average of all the absolute values of the calculated differences between the readings in the dataset. As seen in Table 2.2, the percentage of the

readings within an MAE range is used by BHS to determine a grade for the accuracy of a device (Reinders et al., 2005). ME, STDE and MAE are calculated using the equations below,

$$\mathbf{ME} = \frac{\sum_{k=1}^{n} (\mathbf{MAP_{ref}} - \mathbf{MAP_{ratio}})_k}{n} \quad \text{Equation 4.1}$$

$$\mathbf{STDE} = \sqrt{\frac{\sum_{k=1}^{n} \{(\mathbf{MAP_{ref}} - \mathbf{MAP_{ratio}})_k - \mathbf{ME}\}^2}{n}} \quad \text{Equation 4.2}$$

$$\mathbf{MAE} = \frac{\sum_{k=1}^{n} |(\mathbf{MAP_{ref}} - \mathbf{MAP_{ratio}})_k|}{n} \quad \text{Equation 4.3}$$

where n is the number of recordings, MAP_{ref} corresponds to the reference MAP calculated using Equation 2.4, 2.5 or 2.6 and $(MAP_{ref} - MAP_{ratio})_k$ corresponds to the difference between the reference MAP and MAP_{ratio} for the k^{th} recording.

To assess the variability of the proposed algorithm, the average STandard Deviation (STD_{avg}) is calculated in the following manner,

$$\mathbf{STD_{avg}} = \sqrt{\frac{\sum_{i=1}^{N} \mathbf{STD_i^2}}{N}} \quad \text{Equation 4.4}$$

where N is the number of subjects and STD_i corresponds to STD of the MAP values calculated for the i^{th} subject. The variability of the estimates of the different approaches can be compared by comparing the different STD_{avg} calculated for each approach.

4.1 Dataset from Healthy Subjects

The accuracy of the Omron HEM790IT monitor used with the dataset of healthy subjects has already been validated in healthy subjects (Coleman et al., 2008), and thus provides a reasonable reference MAP values to compare MAP_{ratio}, obtained using the proposed algorithm, to.

4.1.1 Results

Table 4.1 compares ME, STDE and MAE - between the Omron MAP values calculated using the three formulas and MAP_{ratio} - of the entire dataset of healthy subjects, including the one outlier (discussed below), while Table 4.2 shows the same parameters after the outlier is removed from the dataset. The outlier is one subject, whose calculated ME and MAE values - for all three days of the measurement procedure - were greater than the average ME and MAE of the entire dataset by more than twice the STDE of the dataset.

	$MAP_{33\%}$ - MAP_{ratio}	**$MAP_{40\%}$ - MAP_{ratio}**	**MAP_{HR} - MAP_{ratio}**
ME (mmHg)	- 6.0	- 3.4	- 2.9
STDE (mmHg)	7.6	7.8	7.7
MAE (mmHg)	6.6	5.3	5.2

Table 4.1 ME, STDE and MAE of the differences between the three calculated reference MAP values and MAP_{ratio} for the entire dataset from healthy subjects, including the outlier.

	$MAP_{33\%}$ - MAP_{ratio}	**$MAP_{40\%}$ - MAP_{ratio}**	**MAP_{HR} - MAP_{ratio}**
ME (mmHg)	- 3.8	- 1.2	- 0.8
STDE (mmHg)	4.1	4.3	4.4
MAE (mmHg)	4.5	3.3	3.3

Table 4.2 ME, STDE and MAE of the differences between the three calculated reference MAP values and MAP_{ratio} for the dataset from healthy subjects after excluding the outlier.

Tables 4.3 and 4.4 show the different STD_{avg} values of the dataset of healthy subjects - with and without the outlier, respectively.

	$MAP_{33\%}$	**$MAP_{40\%}$**	**MAP_{HR}**	**MAP_{ratio}**
STD_{avg} (mmHg)	2.2	2.2	2.3	2.9

Table 4.3 Average STD of the reference MAP values and MAP_{ratio} of the entire dataset from healthy subjects, including the outlier.

	$MAP_{33\%}$	$MAP_{40\%}$	MAP_{HR}	MAP_{ratio}
STD_{avg} (mmHg)	2.2	2.3	2.4	3.0

Table 4.4 Average STD of the reference MAP values and MAP_{ratio} of the dataset from healthy subjects after excluding the outlier.

4.1.2 Discussion of the Results with Healthy Subjects

The focus of this analysis will be on Tables 4.2 and 4.4, where the outlier was excluded from the results. It should be noted that even though the individual ME and MAE values of the outlier subject are greater than ME and MAE of the entire dataset by more than twice the STDE of the entire dataset, STD of the outlier's MAP values is close enough to the STD_{avg} of the entire dataset, as its removal had little or no impact on STD_{avg}. This can be observed by comparing Tables 4.3 and 4.4.

The ANSI/AAMI SP10 protocol requires ME of SBP and DBP readings to be within ±5 mmHg and STDE to be within 8 mmHg of the reference readings. However, since MAP is the main parameter under investigation in this study, those criteria are applied to MAP to determine the accuracy of the proposed method. Regardless of which of the three formulas is the best or most accurate for calculating MAP, the ME - calculated by subtracting MAP_{ratio} from MAP calculated using each of the three formulas - is found to be well below ±5 mmHg and the STDE is well below 8 mmHg.

According to the error values in Table 4.2, with normal subjects, the proposed approach achieves a performance accuracy that is well within the requirements of the ANSI/AAMI SP10 protocol when the validated Omron HEM790IT monitor is used as a reference. In addition, with normal subjects, the miniscule differences in Table 4.4 between STD_{avg} of MAP_{ratio} and the

reference ones show that the estimates based on the proposed method have only a slightly higher variability when compared to that of the estimates based on the reference device.

4.2 Dataset from Patients

The accuracy of the BpTRU BPM-200 monitor used with the dataset of patients has already been validated in healthy subjects (Mattu et al., 2004).

4.2.1 Results

Table 4.5 refers to the dataset of patients, where ME, STDE and MAE between the BpTRU MAP values calculated using the three formulas and MAP_{ratio} are compared.

	$MAP_{33\%}$ - MAP_{ratio}	$MAP_{40\%}$ - MAP_{ratio}	MAP_{HR} - MAP_{ratio}
ME (mmHg)	0.0	3.4	4.2
STDE (mmHg)	6.1	6.3	6.5
MAE (mmHg)	4.9	5.6	6.0

Table 4.5 ME, STDE and MAE of the differences between the three calculated reference MAP values and MAP_{ratio} for the dataset from patients.

The four STD_{avg} values calculated using the dataset of patients are shown in Table 4.6.

	$MAP_{33\%}$	$MAP_{40\%}$	MAP_{HR}	MAP_{ratio}
STD_{avg} (mmHg)	4.9	5.0	5.0	2.6

Table 4.6 Average STD of the reference MAP values and MAP_{ratio} of the dataset from patients.

4.2.2 Discussion of the Results with Patients

Given the published validation of the BpTRU monitor (Mattu et al., 2004), it can provide reasonable reference SBP and DBP values in healthy subjects. However, for cardiovascular patients in general, and atrial fibrillation patients in particular, this may not be the case.

Therefore, calculating MAP from inaccurate SBP and DBP values would further augment inaccuracies.

Table 4.5 shows the differences in the readings between MAP_{ratio}, obtained using the proposed algorithm, and the different MAP values calculated using the three formulas using SBP, DBP and HR readings obtained using the BpTRU monitor. Even though the ME calculated by subtracting MAP_{ratio} from MAP calculated using each of the three formulas is found to be lower than ±5 mmHg and the STDE lower than 8 mmHg, not much can be said about the accuracy of the proposed method because the accuracy of the reference monitor itself is questionable for these patients.

Since accuracy is not a reliable measure of the performance of the proposed algorithm with the patient data, variability is focused on as the evaluation criterion. Table 4.6 compares the STD_{avg} of MAP calculated using the three formulas from SBP, DBP and HR readings obtained using the BpTRU monitor and MAP_{ratio}, that is obtained using the proposed algorithm. The significantly lower STD_{avg} of MAP_{ratio} - almost half of STD_{avg} of the reference MAP values - means that the variability of the blood pressure estimates, obtained using the proposed approach, is substantially lower than that obtained with BpTRU. This leads to the conclusion that, in atrial fibrillation, the proposed algorithm has substantially higher precision than the validated BpTRU BPM-200 monitor.

4.3 Why Ratio = 2 ?

Even though Equation 2.3 is not the origin of the ratio value of 2, the 2 can be derived from the classical Equation 2.3 in the following manner:

$$\mathbf{MAP} = \mathbf{A} * \mathbf{SBP} + \mathbf{B} * \mathbf{DBP} \qquad \text{Equation 4.5}$$

where the ratio of B to A is the ratio value where the cuff pressure is equal to MAP. Table 4.7 shows the different ratio values determined using the reference Equations 2.4, 2.5 and 2.6.

	A	B	Ratio
33% Formula	0.3333	0.6667	0.667 ÷ 0.333 = 2
40% Formula	0.4	0.6	0.6 ÷ 0.4 = 1.5
HR Formula	0.3333 + (0.0012HR)	0.6667 - (0.0012HR)	Varies depending on HR

Table 4.7 The different coefficients A and B from Equation 4.5 and their corresponding ratio value for the three reference formulas - Equations 2.4, 2.5 and 2.6.

The different ratio values were tested to determine which one would produce the most accurate estimates of MAP, by comparing the MAP estimated using the proposed algorithm but with the different ratio values (2, 1.5 and the HR-dependant ratio) against the corresponding reference MAP. Table 4.8 summarizes the results of the comparisons using the dataset of healthy subjects, where Ratio = 2 is the only reasonable value since the other ratio values consistently under-estimated MAP with a large error margin.

	Ratio = 2	Ratio = 1.5	HR Ratio
ME (mmHg)	- 3.8	-20.2	-21.5
STDE (mmHg)	4.1	4.5	4.0
MAE (mmHg)	4.5	20.2	21.5

Table 4.8 ME, STDE and MAE between MAP_{ratio} and the reference MAP. The first column compares MAP_{ratio} where ratio = 2 and its corresponding $MAP_{33\%}$. The second column compares MAP_{ratio} where ratio = 1.5 and its corresponding $MAP_{40\%}$. The last column compares MAP_{ratio} where the ratio is a function of HR and its corresponding MAP_{HR}.

4.4 Exploring Representative Cases

In this section, three cases are analyzed and discussed. The first case is a recording from a healthy subject, the second and third are from atrial fibrillation patients.

4.4.1 Case 1

Case 1 shows how the proposed algorithm functions in general. Figure 3.1 shows the cuff pressure waveform recorded from a healthy subject. After detrending and filtering the cuff pressure waveform, the result is the oscillometric waveform in Figure 3.2. Next, the peaks and troughs of the oscillometric waveform are identified and located, as seen in Figure 3.5. Finally, a ratio value is calculated for every identified pulse (from its peak and trough) to produce the array of ratio values. All the ratio values in the array are subtracted by 2 and the first zero-crossing is located - in this case the first zero-crossing is highlighted by the red circle in Figure 3.7. The cuff pressure at that point (around 88.6 mmHg) is determined to be MAP.

4.4.2 Case 2

Case 2 serves to demonstrate how the proposed algorithm deals with atrial fibrillation. Figure 4.1 shows the cuff pressure waveform recorded from an atrial fibrillation patient.

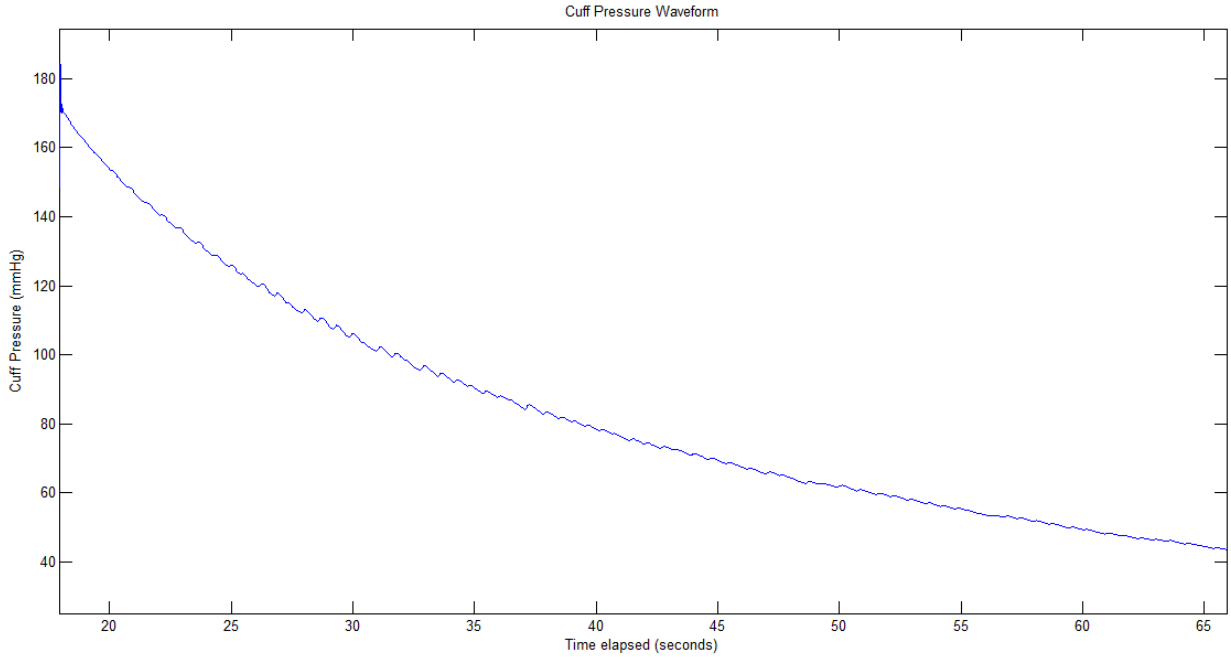


Figure 4.1 Cuff pressure waveform obtained from the atrial fibrillation patient in Case 2.

Detrending and filtering the cuff pressure waveform produces the oscillometric waveform in Figure 4.2.

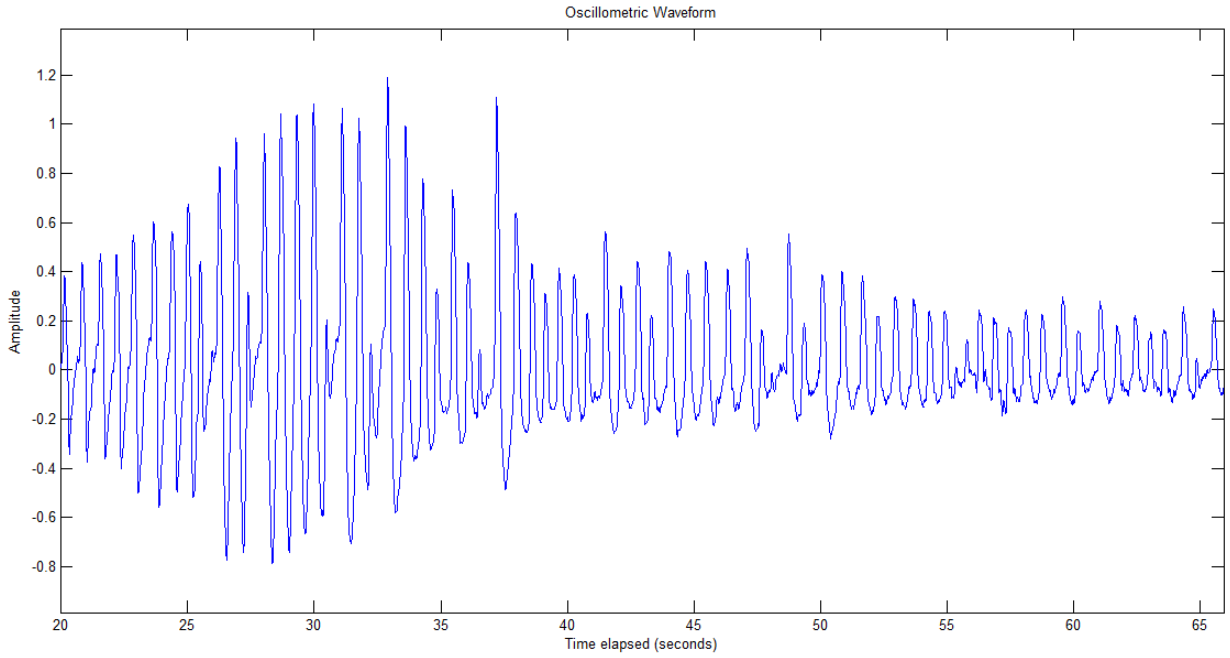


Figure 4.2 Oscillometric waveform of the atrial fibrillation patient in Case 2.

Next, a peak detection algorithm is used to find and locate the peaks and troughs on the oscillometric waveform. A peak detection algorithm would normally locate all the peaks and troughs, such as in Figure 4.3. That is not desirable in atrial fibrillation patients because the premature pulses may lead to erroneous estimation of MAP. Therefore, the premature pulses are discarded - where a pulse is identified as a premature pulse if its peak amplitude from the zero line is lower than the 3-point moving average by more than 0.2*STD. Figure 4.4 shows the result of the peak detection algorithm after the premature pulses are discarded.

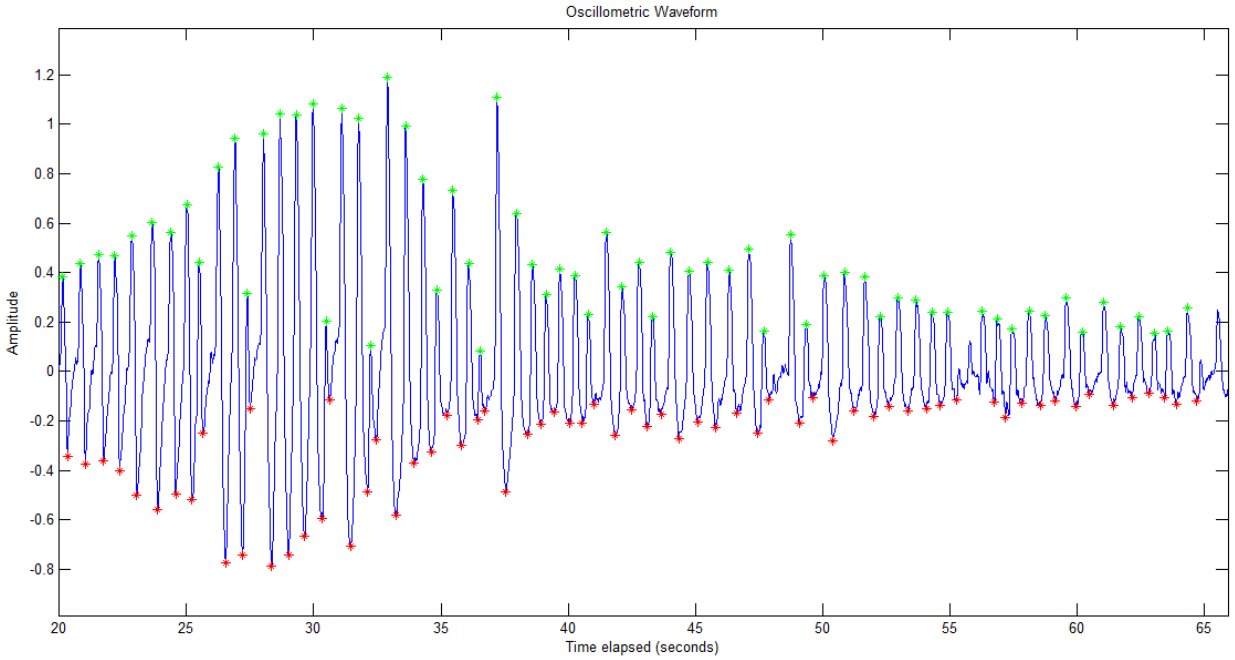


Figure 4.3 Result of a typical peak detection algorithm on the oscillometric waveform of the atrial fibrillation patient in Case 2. Green markings indicate peaks and red markings indicate troughs.

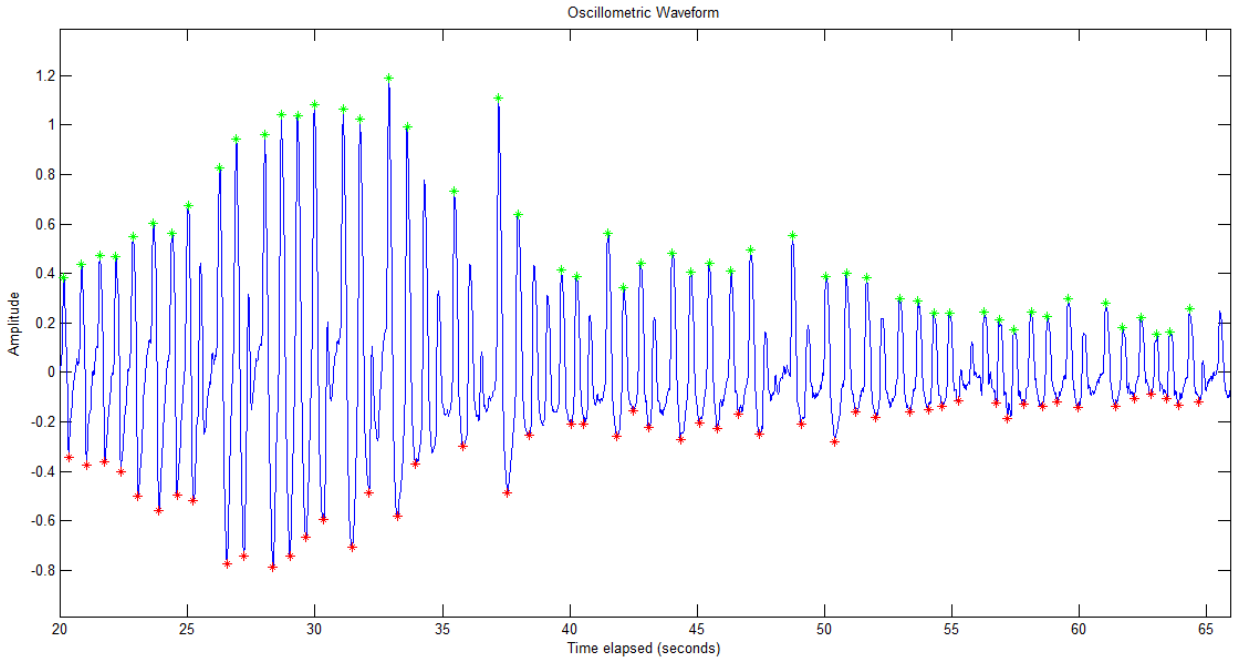


Figure 4.4 Result of the peak detection algorithm in the proposed work on the oscillometric waveform of the atrial fibrillation patient in Case 2. Premature pulses are discarded and the peaks and troughs left are marked in green and red, respectively.

Finally, the ratio array is calculated and subtracted by 2 to determine the first zero-crossing, which in turns leads to the estimation of MAP. Figure 4.5 shows the ratio plot of the atrial fibrillation patient if no pulses are disregarded and Figure 4.6 shows the same plot but after the premature pulses are discarded.

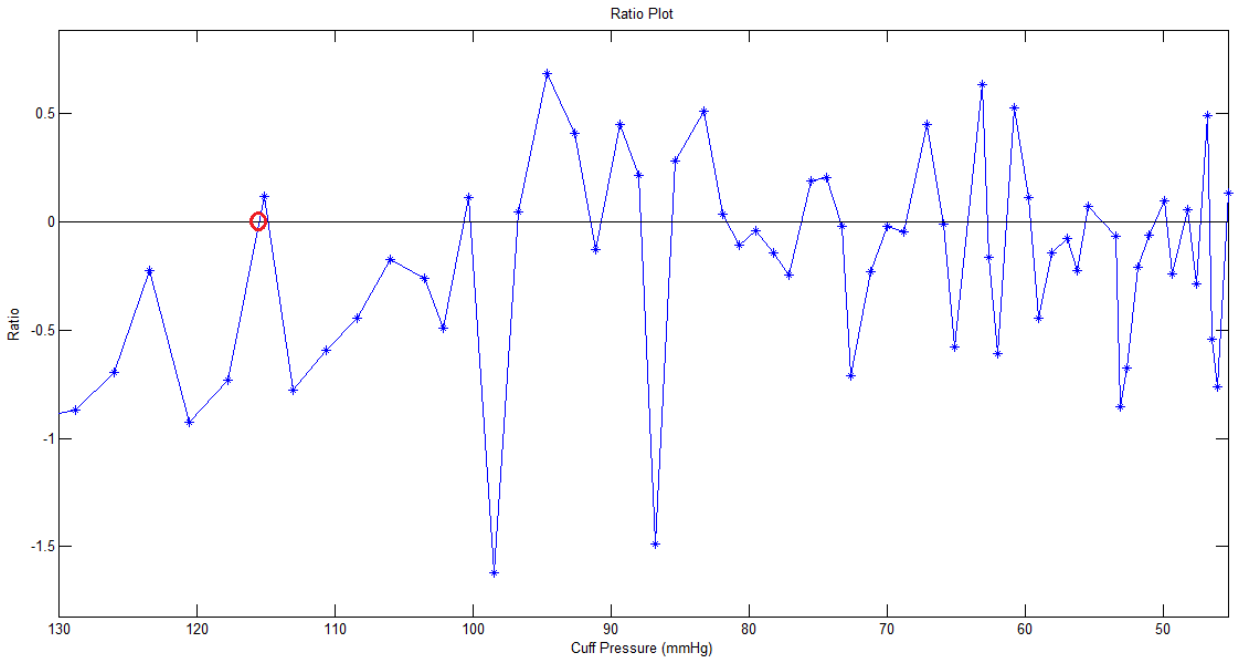


Figure 4.5 Plot of the ratio values of the atrial fibrillation patient in Case 2 before discarding any premature pulses. The red circle indicates the first zero-crossing.

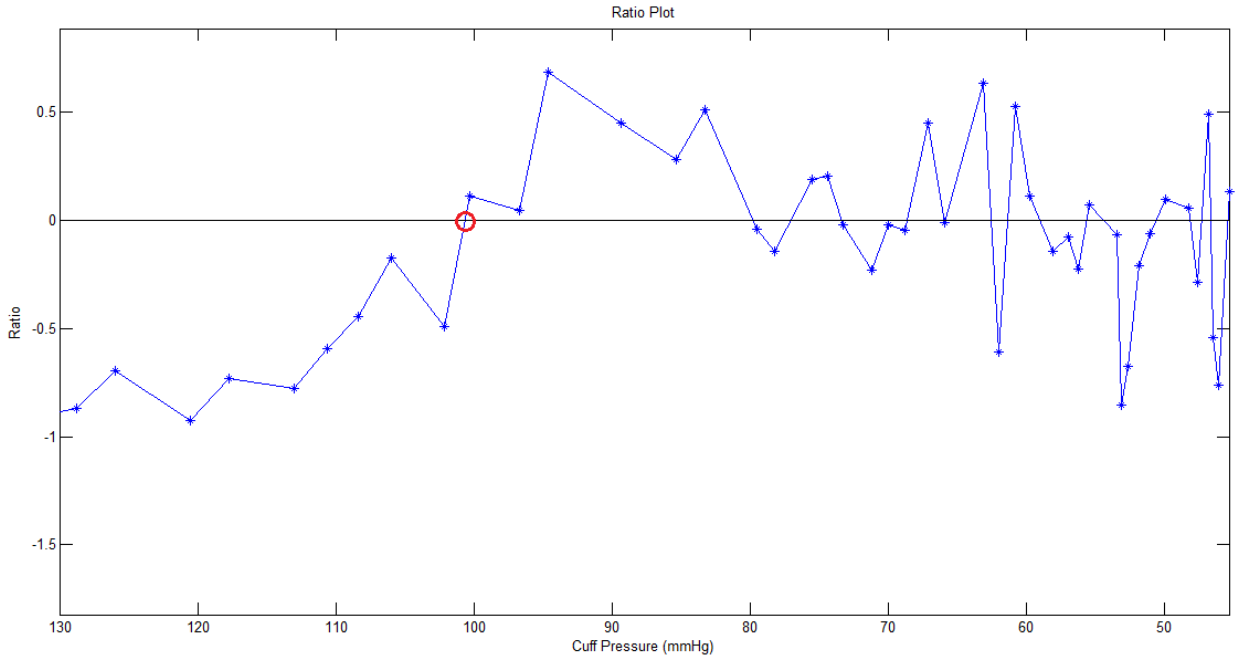


Figure 4.6 Plot of the ratio values of the atrial fibrillation patient in Case 2 after the premature pulses are discarded. The red circle indicates the first zero-crossing.

Since MAP is determined as the cuff pressure where the first zero-crossing occurs, Figure 4.5 points that MAP is around 115 mmHg whereas Figure 4.6 points that MAP is around 100 mmHg. The higher MAP estimate from Figure 4.5 is due to the early zero-crossing occurring due to the noisy ratio plot, where the premature pulses are included in the analysis.

4.4.3 Case 3

As mentioned in Chapter 3, out of both datasets (from patients and healthy subjects), there are five cases where the ratio never reaches 2. Those five cases came from two atrial fibrillation patients. Case 3 is one of those cases and it demonstrates how the algorithm functions when no zero-crossings are found. Figure 4.7 shows the cuff pressure waveform recorded from an atrial fibrillation patient.

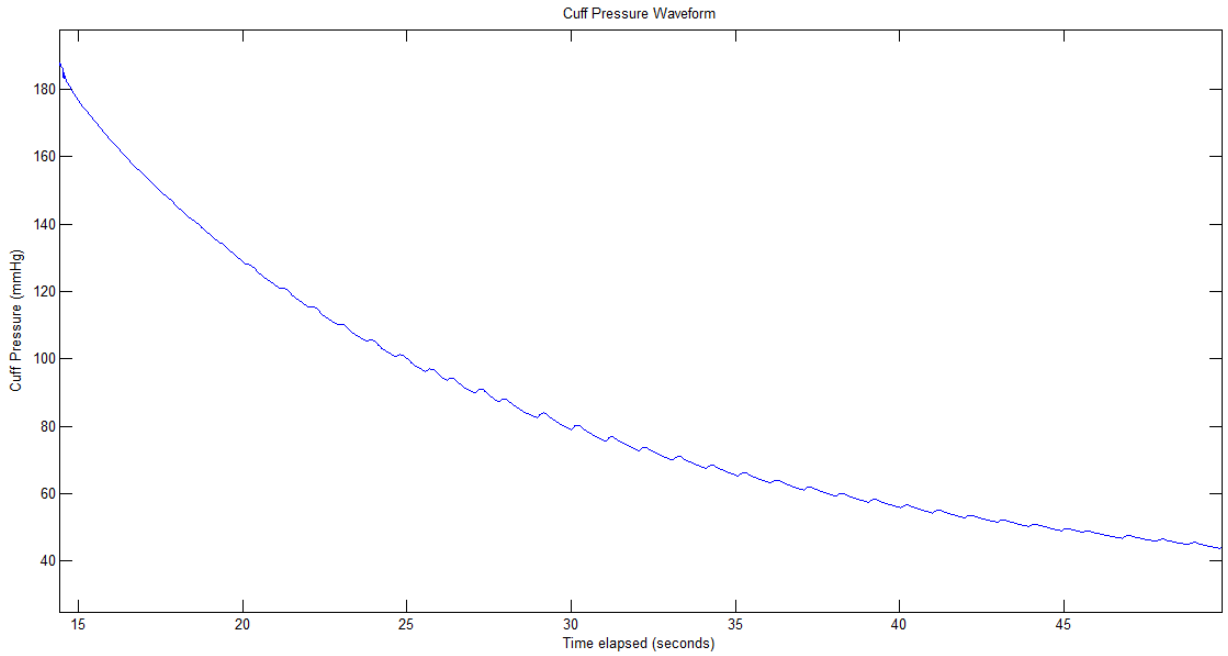


Figure 4.7 Cuff pressure waveform obtained from the atrial fibrillation patient in Case 3.

The oscillometric waveform in Figure 4.8 is the result of detrending and filtering the cuff pressure waveform (Figure 4.7) of the atrial fibrillation patient.

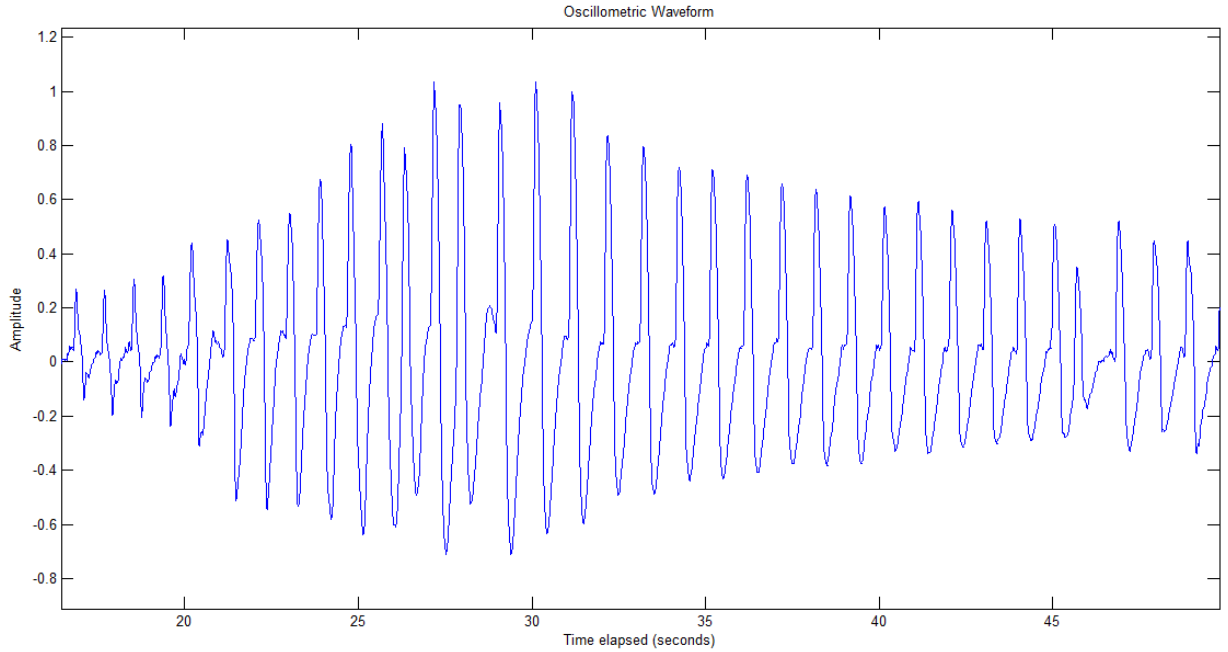


Figure 4.8 Oscillometric waveform of the atrial fibrillation patient in Case 3.

Next, the peaks and troughs of the oscillometric waveform are identified and located as seen in Figure 4.9.

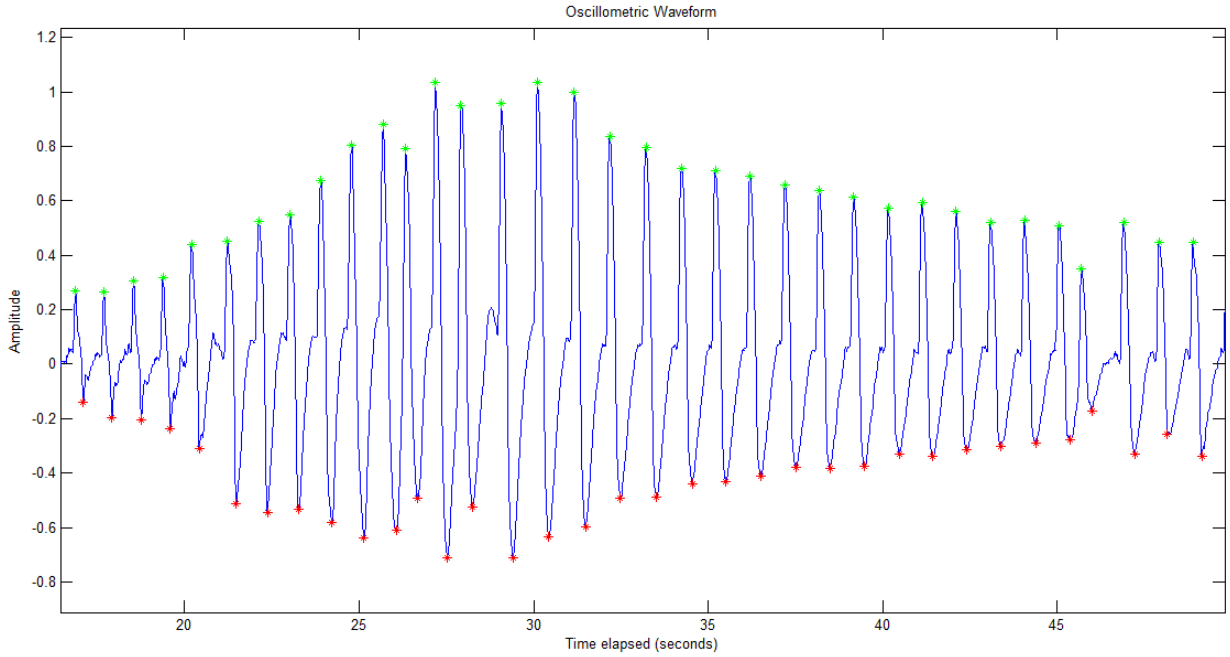


Figure 4.9 Oscillometric waveform of the atrial fibrillation patient in case 3 after applying the peak detection algorithm. The green markings indicate the peaks and the red markings indicate the troughs.

Finally, the ratio array is calculated and subtracted by 2 to determine the first zero-crossing. If no zero-crossing are found, such as in Figure 4.10, this means that the ratio values never reached 2. In cases such as this, the cuff pressure where the ratio is closest to the zero line is determined to be equal to MAP. In Figure 4.10, the point closest to the zero line occurs at a cuff pressure around 88 mmHg. Hence MAP is estimated to be 88 mmHg.

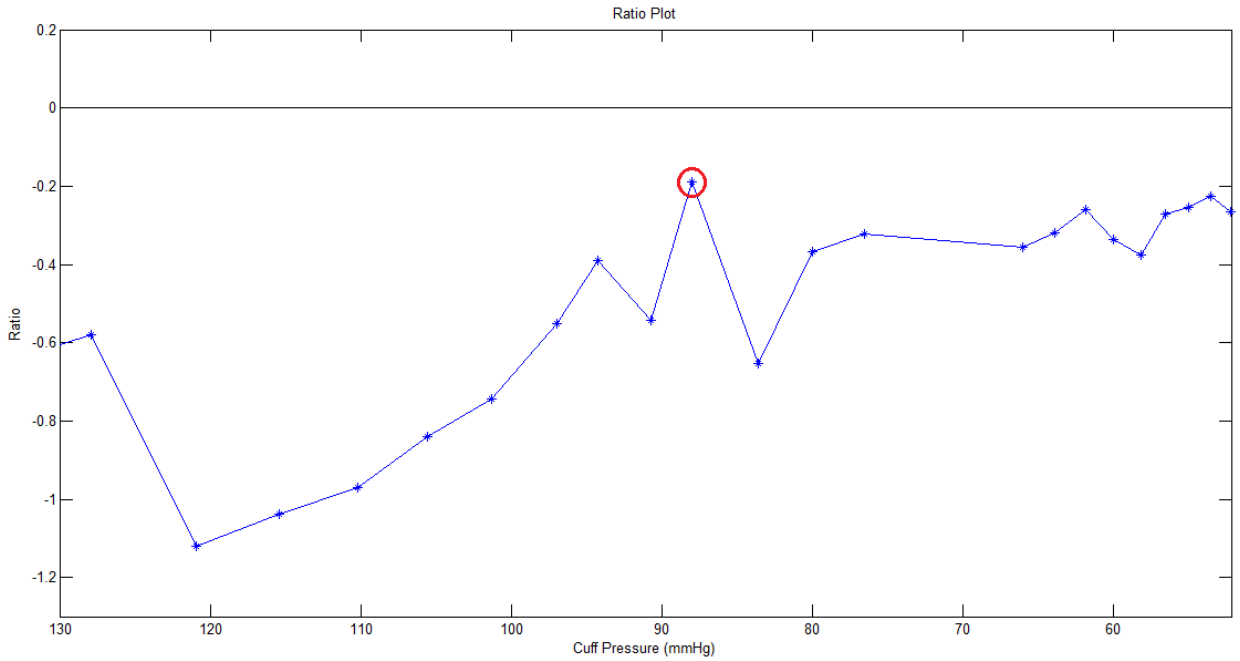


Figure 4.10 Plot of the ratio values of the atrial fibrillation patient in Case 3. No zero-crossings are found so the point closest to the zero line, indicated by the red circle, is used to determine MAP.

Chapter 5. Conclusion

5.1 Summary of thesis

Blood pressure is a vital sign of significant importance as it provides physiological information about the cardiovascular system and can be an indicator of a human's well-being and a predictor of diseases. Blood pressure fluctuates continuously over time between a maximum, Systolic Blood Pressure (SBP), and a minimum, Diastolic Blood Pressure (DBP). Atrial fibrillation, a commonly encountered arrhythmia, causes an abnormality in the rhythmic beats of the heart and augments the short-term variability of blood pressure (Jani et al., 2006). As a result, the maximum and minimum levels of blood pressure (SBP and DBP, respectively) fluctuate rapidly. This raises doubts in terms of the accuracy of existing automated blood pressure monitors and the repeatability of the measurements. In the past, studies have pointed out this issue and suggested no solution other than repeating several measurements and calculating the average SBP and DBP (Sykes et al., 1990; Stewartet al., 1995; Kaliujnaya and Kalyuzhny, 2005; Jani et al., 2006; Anastas et al., 2008; Lamb et al., 2010; Stergiou et al., 2012). The current work is intended to address the problem at its root by offering a new approach that provides a more precise estimation of Mean Arterial Pressure (MAP) - the parameter that is typically measured first and then used to calculate SBP and DBP by automated blood pressure monitors - over a single measurement interval.

MAP is also an important hemodynamic parameter on its own for the following reasons:

- MAP reflects the blood pressure fluctuations more accurately than SBP or DBP (Smulyan et al., 2008).

- MAP is monitored when medication is used to control perfusion blood pressure (Razminia et al., 2004).
- MAP is used to calibrate central blood pressure waveforms, such as the ones obtained by tonometry (Vos et al., 2013).
- MAP is used to calculate peripheral vascular resistance (Graf et al., 2010).

Even though automated blood pressure monitors measure MAP and calculate SBP and DBP, most only display SBP and DBP. For comparison purposes, the reference MAP had to be calculated from SBP, DBP and Heart Rate (HR) obtained using reference automated blood pressure monitors. Three equations (Equations 2.4, 2.5 and 2.6) were used to calculate the reference MAP values. A fourth MAP, MAP_{ratio}, was determined using the method proposed in this work by calculating a ratio for each heartbeat and determining the cuff pressure when the ratio reached a value of 2. The proposed method was applied to waveforms recorded using another automated blood pressure monitor.

The performance of the proposed algorithm was assessed by calculating the differences between the reference MAP readings and the proposed MAP readings - individually and as Mean Error (ME) ± STandard Deviation of Error (STDE), in addition to Mean of Absolute Errors (MAE). Average STandard Deviation (STD_{avg}) of each of the three reference MAP readings as well as the proposed MAP was also calculated in order to assess the variability of the readings.

When the Omron monitor is used as the reference for the dataset of healthy subjects, the proposed algorithm achieves a performance accuracy that is well within the ANSI/AAMI SP10 protocol requirements, as well as a similar variability of the readings. For the dataset of patients (77% of whom had atrial fibrillation), the BpTRU monitor is used as the reference. In this case

too, the proposed algorithm achieves a performance accuracy that is well within the requirements of the ANSI/AAMI SP10 protocol. The presence of atrial fibrillation increased the variability of the reference readings, as expected. However, the proposed algorithm performed better by achieving substantially lower variability in the readings than the reference device.

5. 2 Contributions

The following are the main contributions of this thesis:

- Developing a novel algorithm for more precise estimation of MAP.
- Testing and confirming the accuracy of the proposed approach on a dataset of healthy subjects and on a dataset of patients with atrial fibrillation and/or hypertension.
- Contributing to the development of improved non-invasive blood pressure technology for difficult to measure patients.

5.3 Limitations

The ANSI/AAMI SP10 non-invasive measurement standard recommends a one minute delay between every blood pressure measurement procedure (Association for the Advancement of Medical Instrumentation, 2003). The effect of the time gap (waiting) between each measurement was not investigated. Blood pressure fluctuates continuously and separate measurements from the same device could produce different estimations when separated by even a few tens of seconds (Pagani et al., 1986). The subjects' blood pressure could have changed during the wait between the measurements and therefore, that time gap may be a source of error. The best way to compare the algorithms would be by performing both measurements simultaneously to limit the effects of the continuous blood pressure fluctuations. Unfortunately, this was not possible due to concerns over the comfort, and potentially safety, of the patients.

Even though the invasive intra-arterial method is considered to be the gold standard for blood pressure measurement, the invasive nature of the procedure limits its use in clinics and homes. The manual auscultatory method continues to be the most common method of clinical blood pressure measurement and should therefore be used to obtain the reference measurements (Pickering et al., 2005). In this thesis, for the dataset of patients, a BpTRU monitor was used to obtain the reference measurements, as it uses the average of six measurements to increase the accuracy of the blood pressure estimations (BpTRU Operator's Manual, 2012), rather than a manual sphygmomanometer. Possible inaccuracies in the BpTRU monitor's blood pressure estimates would be a source of error.

Typically, $MAP_{33\%}$, $MAP_{40\%}$ and MAP_{HR} are calculated using SBP and DBP measured by auscultation (Raamat et al., 2013). In this study, the SBP and DBP used to calculate the reference MAP values are obtained by oscillometry using the reference monitors.

5.4 Future Work

The limitations faced could be addressed in future work. A more definite validation of the proposed algorithm can be investigated by comparing against the invasive gold standard - continuous intra-arterial blood pressure recordings.

The proposed algorithm has been tested on a limited number of subjects and could be tested on larger scale, while focusing on different cardiovascular conditions, for further validation.

The ratio value of 2 could be further investigated in the future to determine exactly why it is so and whether there is an optimum alternate value that provides more accurate MAP estimates. The ratio value of 2 may differ when a different type of cuff is used, and that may be

grounds for future investigation. This does not call into question the proposed approach itself, but it just implies that further validation using different hardware would probably be informative.

MAP is the parameter estimated in this thesis, since it is the main parameter that automated monitors measure and use to calculate SBP and DBP. Future work should include developing new approaches to calculate SBP and DBP, perhaps by combining or fusing the proposed algorithm with the MAA technique to produce more accurate SBP, DBP and MAP estimates, particularly with difficult to measure patients.

Appendix A - Informed Consent Form

Informed Consent Form
ECG-Assisted Blood Pressure Measurement

This research project is being conducted by Professors Miodrag Bolic, Hilmi R. Dajani, and Voicu Z. Groza at the School of Electrical Engineering and Computer Science (SEECS) of the University of Ottawa. [Tel. (613) 562-5800, ext. 6224].

Purpose

The purpose of this study is to develop a novel technology for robust and accurate measurement of blood pressure (BP) in humans. This informed consent form is to make sure that you understand the nature of your involvement in this study, and to obtain your informed consent to participate in this study.

Procedure

For data collection, you will need to either come to our Biomedical Engineering Laboratory in Room 3013, SITE Building, University of Ottawa, 800 King Edward Avenue, Ottawa, ON, K1N 6N5 or to the Ottawa Cardiovascular Centre, Suite 502, 1355 Bank Street, Ottawa, ON, K1H 8K7. Data will be collected on a single day and the entire data collection protocol will last about one hour.

We will use noninvasive and unobtrusive physiology monitors to collect electrocardiogram (ECG), BP, and blood oxygen saturation (SpO2) data from you. We will attach one upper arm BP cuff from our device to one of your arms and another cuff from a commercial device to your other arm. Additionally, we will attach one ECG wrist clip from our device to your wrist and a finger clip SpO2 sensor to your index finger. The BP cuff from our device will have embedded dry flexible electrodes for ECG data acquisition. The ECG wrist clip from our device will also have embedded dry electrodes for ECG data acquisition. As an alternative to the ECG wrist clip, you can also touch dry electrodes provided on our device box for ECG data acquisition. The second BP cuff will be a standard upper arm cuff from either of these two commercial BP monitors: BPTru or Omron. To take a measurement, the cuff will be inflated to a pressure above your expected systolic pressure and then deflated slowly until pressure below your diastolic pressure is reached. One measurement will be completed within 30-70 seconds. We will alternately take twelve measurements from your right arm and twelve from your left arm, resulting in a total of twenty four measurements. During data collection, you will be seated comfortably in a chair and will be asked to stay calm and breathe normally.

Subject Description

Healthy adults (>18y) as well as adult patients with conditions namely hypertension, AF, obesity, and AS will be recruited for the study. For healthy subjects, the inclusion criterion will be individuals who do not suffer from any of the following conditions: hypertension, atrial fibrillation (AF), obesity (Body Mass Index: BMI>29.9), & atherosclerosis (AS), and who do not take any medications for these conditions. For patients, the inclusion criterion will be individuals who suffer from one or more of the following conditions: hypertension, AF, obesity (BMI>29.9), & AS, for which they may or may not be on medication.

To determine inclusion or exclusion, you will be asked to fill a questionnaire whereby the following information will be collected:

1. ***Personal Information:*** Your name, gender, date of birth, height, and weight.

2. ***Medical Information:*** Whether you have any of the following conditions – hypertension, atrial fibrillation (AF), stroke, transient ischemic attack (TIA), angina, heart attack, angioplasty or bypass, claudication (blockage of peripheral arteries), peripheral angioplasty of bypass, amputation due to poor blood supply, and abdominal aortic aneurysm (current or repaired).

Risks To Participants

There is no danger or risk to health associated with this study. All procedures have been pre-tested and they have been used routinely for many years in hospitals, clinics, and laboratories. There may be minor physical discomfort in the upper arm due to cuff inflation and deflation. However, this discomfort is not serious and goes away as soon as pressure in the cuff decreases.

Nonetheless, participants are free to ask the researchers to stop the test at any time.

Compensation

For participating in the study, you will be paid a token allowance of $25 to cover for your travel, parking, and time.

Confidentiality

Any information about you collected during the study will be kept strictly confidential. Your name will not be associated with the collected data in any way. While the results may appear in student's dissertations and may also be published, you will not be identified.

The data collection will be conducted by Dr. Bolic's research associate Dr. Saif Ahmad. The records will be kept on computer files. The files will be password-protected. The data will be kept in Dr. Bolic's office when not in use and will be conserved for a maximum period of 15 years, after which the computer files will be deleted and the consent forms will be shredded.

In Closing

With your participation, you will be given a copy of this consent form. At the conclusion of the study, should you wish, you will be provided with a summary of the results. However, these results cannot be used for any medical diagnosis. If you need to see your medical doctor about your results, you are advised to do so. You may ask questions at any time, even after signing this consent form.

Withdrawing From The Study

Your participation in this study is voluntaryand it will have no bearing on relationships such as patient-doctor, student-professor, family-family, and friend-friend. For example, if you are a patient at the Clinic, participation or non-participation in the research will have no effect on the treatment or services that you are receiving at the Clinic.

You may withdraw from the study at any time, by verbally informing the investigator or any of the researchers, even after signing the consent form or even after data collection has started. There will be no consequences following this action.

If you enroll for the study, show up on the day of the study, and start the data collection protocol, you will be compensated even if you withdraw from the study before completion of data collection.

If you have any concerns with regards to the ethical conduct of the study, you may contact the Protocol Officer for Ethics in Research, University of Ottawa, Tabaret Hall, 550 Cumberland Street, Room 154, Ottawa, ON K1N 6N5, tel. (613) 562-5387, email: ethics@uottawa.ca.

Signatures

I have read the above description of the study and understand the conditions of participation. My signature indicates that I agree to participate in the study.

Would you like to receive a summary of your results: YES / NO

Name of Participant (*Please Print*):

_____

Participant's Phone Number:

_____

Participant's Email: _____

Participant's Signature: _____

Date: _____

Researcher's Signature: _____

Date: _____

Appendix B - Participant Questionnaire

ECG-Assisted Blood Pressure Measurement

Participant's Name:_____ Date:_____

Gender: Male / Female Date of Birth:_____ Height:_____ Weight:_____

Please tick (✓) against the YES/NO box below as appropriate:

SNO	Question	YES	NO	If YES, please provide details (medications, etc.)
1	Do you suffer from hypertension?			
2	Do you suffer from atrial fibrillation (AF)?			
3	Do you suffer from arterial stiffness or atherosclerosis? Some manifestations are listed below. Please tick (✓) where appropriate:			
3-a	Stroke?			
3-b	Transient ischemic attack (TIA) ?			
3-c	Angina?			
3-d	Heart attack?			
3-e	Angioplasty or bypass?			

3-f	Claudication (blockage of peripheral arteries) ?			
3-g	Peripheral angioplasty or bypass?			
3-h	Amputation due to poor blood supply?			
3-i	Abdominal aortic aneurysm (current or repaired) ?			

Participant's Signature:_____

For Researchers' Use Only

BMI (Weight/Height^2)		
Obese (BMI>29.9, Circle One)?	**Yes**	**No**

Appendix C - Participant Recruitment Poster

Université d' University of Ottawa

ECG-Assisted Blood Pressure Measurement

We invite you to participate in a blood pressure (BP) measurement study being conducted by a group of researchers at the University of Ottawa. Hypertension is an extremely common disease that affects about one billion people worldwide. Patients with accompanying chronic conditions such as atrial fibrillation, obesity, and arterial stiffness, are at high risk of medical complications including myocardial infarctions and strokes, if their blood pressures are not monitored accurately. Research has shown that current BP monitoring technology fails to provide accurate estimates for the above patient population thus putting them at high risk of developing life-threatening conditions.

The purpose of this study is to test a new system that uses an electrocardiogram (ECG) signal to increase accuracy of BP measurement. By monitoring BP more accurately, the proposed system promises to reduce the risk of dangerous medical consequences and save healthcare dollars.

All physiological data for this study will be collected using non-invasive/unobtrusive monitoring devices. More specifically, we will collect ECG, BP, and blood oxygen saturation (SpO2) data. ECG data will be collected using innovative dry electrode technology whereby the electrodes are embedded in an upper arm BP cuff. BP data will be collected using standard upper arm cuffs while SpO2 data will be collected using a finger clip sensor. The knowledge gained from the collection and analysis of this data is expected to help in the development of a better and more accurate BP monitoring device.

Healthy adults (>18 y) as well as adult **patients** are invited to participate in the study.

Healthy Adults: Adults who do not suffer from any of the following conditions: hypertension, atrial fibrillation, obesity (Body Mass Index: BMI>29.9), & arterial stiffness, and who do not take any medications for these conditions can participate in this study.Please note that your participation in this study is purely voluntary and it will have no bearing on relationships such as student-professor, family-family, and friend-friend.

Adult Patients: Adults who suffer from one or more of the following conditions: hypertension, AF, obesity (BMI>29.9), & AS, for which they may or may not be on medication can participate in this study.Please note that your participation in this study is purely voluntary. Participation or non-participation in the research will have no effect on the treatment or services that you are receiving at the Clinic.

Compensation: For your participation, you will be paid a token allowance of $25 for travel, parking, and your time.

Data collection will take place on a single day either in our Biomedical Engineering Laboratory, Room 3013, SITE Building, University of Ottawa, 800 King Edward Avenue, Ottawa, ON, K1N 6N5 or at the Ottawa Cardiovascular Centre, Suite 502, 1355 Bank Street, Ottawa, ON, K1H 8K7, and will last about one hour. You will be seated comfortably in a chair and asked to be calm and breathe normally. There are no risks associated with the study or the measurement devices.

Participants for this study will be selected on a first come/first served basis.

If you are interested in being part of our study or have any other questions, please contact the investigator (s), Dr. Miodrag Bolic at (613) 562-5800, ext. 6224, mbolic@site.uottawa.caor Dr. Saif Ahmad at 613-884-4094, sahmad@site.uottawa.ca.

Your collaboration is appreciated.

Miodrag Bolic, Ph.D

Appendix D - Experimental Results with Dataset of Healthy Subjects

Each table block refers to one subject during one day. S01D1 refers to subject 1 on day 1. The first column shows the subject ID, day number and recording number. The next three columns show the values displayed by the Omron monitor and the next three columns show the three MAP values calculated from those parameters. The next column shows the MAP estimated using the proposed approach and the last three columns show the difference between the reference MAP values and MAP estimated using the proposed approach.

S01 D1	Omron Recordings			Omron Calculated MAP			MAP_{ratio}	Errors (Reference - MAP_{ratio})			
Rec #	SBP	DBP	HR	$MAP_{33\%}$	$MAP_{40\%}$	MAP_{HR}		$MAP_{33\%}$	$MAP_{40\%}$	MAP_{HR}	
1	136.0	84.0	70.0	101.2	104.8	105.5	103.4	-2.2	1.4	2.1	
2	119.0	79.0	65.0	92.2	95.0	95.3	97.0	-4.8	-2.0	-1.7	
3	132.0	76.0	67.0	94.5	98.4	99.0	99.8	-5.3	-1.4	-0.8	
Mean	129.0	79.7	67.3	95.9	99.4	99.9	100.1	-4.1	-0.7	-0.1	ME
STD	8.9	4.0	2.5	4.7	5.0	5.2	3.2	1.6	1.8	2.0	STDE
								4.1	1.6	1.5	MAE

S01 D2	Omron Recordings			Omron Calculated MAP			MAP_{ratio}	Errors (Reference - MAP_{ratio})			
Rec #	SBP	DBP	HR	$MAP_{33\%}$	$MAP_{40\%}$	MAP_{HR}		$MAP_{33\%}$	$MAP_{40\%}$	MAP_{HR}	
1	131.0	80.0	63.0	96.8	100.4	100.7	100.9	-4.1	-0.5	-0.2	
2	127.0	79.0	61.0	94.8	98.2	98.4	100.1	-5.3	-1.9	-1.7	
3	119.0	79.0	64.0	92.2	95.0	95.3	96.2	-4.0	-1.2	-0.9	
Mean	125.7	79.3	62.7	94.6	97.9	98.1	99.1	-4.4	-1.2	-1.0	ME
STD	6.1	0.6	1.5	2.3	2.7	2.7	2.5	0.7	0.7	0.8	STDE
								4.4	1.2	1.0	MAE

S01 D3	Omron Recordings			Omron Calculated MAP			MAP_{ratio}	Errors (Reference - MAP_{ratio})			
Rec #	SBP	DBP	HR	$MAP_{33\%}$	$MAP_{40\%}$	MAP_{HR}		$MAP_{33\%}$	$MAP_{40\%}$	MAP_{HR}	
1	117.0	78.0	64.0	90.9	93.6	93.9	89.0	1.9	4.6	4.9	
2	115.0	75.0	64.0	88.2	91.0	91.3	88.4	-0.2	2.6	2.9	
3	114.0	74.0	61.0	87.2	90.0	90.1	89.7	-2.5	0.3	0.4	
Mean	115.3	75.7	63.0	88.8	91.5	91.8	89.0	-0.3	2.5	2.7	ME
STD	1.5	2.1	1.7	1.9	1.9	1.9	0.7	2.2	2.2	2.2	STDE
								1.5	2.5	2.7	MAE

S02 D1	Omron Recordings			Omron Calculated MAP			MAP_{ratio}	Errors (Reference - MAP_{ratio})			
Rec #	SBP	DBP	HR	$MAP_{33\%}$	$MAP_{40\%}$	MAP_{HR}		$MAP_{33\%}$	$MAP_{40\%}$	MAP_{HR}	
1	109.0	72.0	71.0	84.2	86.8	87.4	88.6	-4.4	-1.8	-1.2	
2	106.0	69.0	68.0	81.2	83.8	84.2	86.3	-5.1	-2.5	-2.1	
3	108.0	71.0	68.0	83.2	85.8	86.2	88.5	-5.3	-2.7	-2.3	
Mean	107.7	70.7	69.0	82.9	85.5	85.9	87.8	-4.9	-2.3	-1.9	ME
STD	1.5	1.5	1.7	1.5	1.5	1.6	1.3	0.5	0.5	0.5	STDE
								4.9	2.3	1.9	MAE

S02 D2	Omron Recordings			Omron Calculated MAP			MAP_{ratio}	Errors (Reference - MAP_{ratio})			
Rec #	SBP	DBP	HR	$MAP_{33\%}$	$MAP_{40\%}$	MAP_{HR}		$MAP_{33\%}$	$MAP_{40\%}$	MAP_{HR}	
1	118.0	76.0	75.0	89.9	92.8	93.6	93.3	-3.4	-0.5	0.3	
2	115.0	73.0	75.0	86.9	89.8	90.6	93.6	-6.7	-3.8	-3.0	
3	118.0	73.0	72.0	87.9	91.0	91.7	95.6	-7.8	-4.6	-3.9	
Mean	117.0	74.0	74.0	88.2	91.2	92.0	94.2	-7.2	-4.2	-3.4	ME
STD	1.7	1.7	1.7	1.5	1.5	1.5	1.3	0.7	0.6	0.6	STDE
								6.0	3.0	2.4	MAE

S02 D3	Omron Recordings			Omron Calculated MAP			MAP_{ratio}	Errors (Reference - MAP_{ratio})			
Rec #	SBP	DBP	HR	$MAP_{33\%}$	$MAP_{40\%}$	MAP_{HR}		$MAP_{33\%}$	$MAP_{40\%}$	MAP_{HR}	
1	111.0	73.0	64.0	85.5	88.2	88.5	96.0	-10.5	-7.8	-7.5	
2	117.0	74.0	63.0	88.2	91.2	91.4	94.3	-6.1	-3.1	-2.9	
3	104.0	73.0	67.0	83.2	85.4	85.7	97.4	-14.2	-12.0	-11.7	
Mean	110.7	73.3	64.7	85.7	88.3	88.5	95.9	-10.2	-7.6	-7.4	ME
STD	6.5	0.6	2.1	2.5	2.9	2.9	1.6	4.0	4.5	4.4	STDE
								10.2	7.6	7.4	MAE

S03 D1	Omron Recordings			Omron Calculated MAP			MAP_{ratio}	Errors (Reference - MAP_{ratio})			
Rec #	SBP	DBP	HR	$MAP_{33\%}$	$MAP_{40\%}$	MAP_{HR}		$MAP_{33\%}$	$MAP_{40\%}$	MAP_{HR}	
1	84.0	57.0	61.0	65.9	67.8	67.9	75.9	-10.0	-8.1	-8.0	
2	84.0	55.0	63.0	64.6	66.6	66.8	71.1	-6.5	-4.5	-4.3	
3	82.0	55.0	53.0	63.9	65.8	65.6	73.8	-9.9	-8.0	-8.2	
Mean	83.3	55.7	59.0	64.8	66.7	66.8	73.6	-8.2	-6.3	-6.3	ME
STD	1.2	1.2	5.3	1.0	1.0	1.1	2.4	2.4	2.5	2.7	STDE
								8.8	6.9	6.8	MAE

S03 D2	Omron Recordings			Omron Calculated MAP			MAP_{ratio}	Errors (Reference - MAP_{ratio})			
Rec #	SBP	DBP	HR	$MAP_{33\%}$	$MAP_{40\%}$	MAP_{HR}		$MAP_{33\%}$	$MAP_{40\%}$	MAP_{HR}	
1	80.0	55.0	61.0	63.3	65.0	65.1	71.0	-7.8	-6.0	-5.9	
2	80.0	57.0	62.0	64.6	66.2	66.3	68.7	-4.1	-2.5	-2.4	
3	81.0	57.0	59.0	64.9	66.6	66.6	68.9	-4.0	-2.3	-2.3	
Mean	80.3	56.3	60.7	64.3	65.9	66.0	69.5	-5.3	-3.6	-3.5	ME
STD	0.6	1.2	1.5	0.9	0.8	0.8	1.3	2.1	2.1	2.1	STDE
								5.3	3.6	3.5	MAE

S03 D3	Omron Recordings			Omron Calculated MAP			MAP_{ratio}	Errors (Reference - MAP_{ratio})			
Rec #	SBP	DBP	HR	$MAP_{33\%}$	$MAP_{40\%}$	MAP_{HR}		$MAP_{33\%}$	$MAP_{40\%}$	MAP_{HR}	
1	83.0	61.0	64.0	68.3	69.8	69.9	82.4	-14.1	-12.6	-12.5	
2	86.0	63.0	63.0	70.6	72.2	72.3	85.5	-14.9	-13.3	-13.2	
3	82.0	60.0	54.0	67.3	68.8	68.7	79.3	-12.0	-10.5	-10.6	
Mean	83.7	61.3	60.3	68.7	70.3	70.3	82.4	-13.5	-11.9	-11.9	ME
STD	2.1	1.5	5.5	1.7	1.7	1.8	3.1	2.0	2.0	1.8	STDE
								13.7	12.1	12.1	MAE

S04 D1	Omron Recordings			Omron Calculated MAP			MAP_{ratio}	Errors (Reference - MAP_{ratio})			
Rec #	SBP	DBP	HR	$MAP_{33\%}$	$MAP_{40\%}$	MAP_{HR}		$MAP_{33\%}$	$MAP_{40\%}$	MAP_{HR}	
1	115.0	76.0	63.0	88.9	91.6	91.8	84.9	4.0	6.7	6.9	
2	106.0	76.0	67.0	85.9	88.0	88.3	90.6	-4.7	-2.6	-2.3	
3	110.0	80.0	66.0	89.9	92.0	92.3	89.6	0.3	2.4	2.7	
Mean	110.3	77.3	65.3	88.2	90.5	90.8	88.4	-0.1	2.2	2.4	ME
STD	4.5	2.3	2.1	2.1	2.2	2.2	3.0	4.4	4.7	4.6	STDE
								3.0	3.9	4.0	MAE

S04 D2	Omron Recordings			Omron Calculated MAP			MAP_{ratio}	Errors (Reference - MAP_{ratio})			
Rec #	SBP	DBP	HR	$MAP_{33\%}$	$MAP_{40\%}$	MAP_{HR}		$MAP_{33\%}$	$MAP_{40\%}$	MAP_{HR}	
1	117.0	81.0	70.0	92.9	95.4	95.9	95.5	-2.6	-0.1	0.4	
2	112.0	79.0	71.0	89.9	92.2	92.7	94.4	-4.5	-2.2	-1.7	
3	119.0	79.0	66.0	92.2	95.0	95.4	95.2	-3.0	-0.2	0.2	
Mean	116.0	79.7	69.0	91.7	94.2	94.7	95.0	-3.4	-0.8	-0.4	ME
STD	3.6	1.2	2.6	1.6	1.7	1.7	0.6	1.0	1.2	1.2	STDE
								3.4	0.8	0.8	MAE

S04 D3	Omron Recordings			Omron Calculated MAP			MAP_{ratio}	Errors (Reference - MAP_{ratio})			
Rec #	SBP	DBP	HR	$MAP_{33\%}$	$MAP_{40\%}$	MAP_{HR}		$MAP_{33\%}$	$MAP_{40\%}$	MAP_{HR}	
1	111.0	80.0	70.0	90.2	92.4	92.8	88.0	2.2	4.4	4.8	
2	115.0	76.0	67.0	88.9	91.6	92.0	87.6	1.3	4.0	4.4	
3	111.0	71.0	64.0	84.2	87.0	87.3	87.0	-2.8	0.0	0.3	
Mean	112.3	75.7	67.0	87.8	90.3	90.7	87.5	-0.8	2.0	2.3	ME
STD	2.3	4.5	3.0	3.2	2.9	3.0	0.5	2.9	2.8	2.9	STDE
								2.1	2.8	3.2	MAE

S05 D1	Omron Recordings			Omron Calculated MAP			MAP_{ratio}	Errors (Reference - MAP_{ratio})			
Rec #	SBP	DBP	HR	$MAP_{33\%}$	$MAP_{40\%}$	MAP_{HR}		$MAP_{33\%}$	$MAP_{40\%}$	MAP_{HR}	
1	117.0	68.0	83.0	84.2	87.6	89.1	86.7	-2.5	0.9	2.4	
2	109.0	68.0	77.0	81.5	84.4	85.3	85.2	-3.7	-0.8	0.1	
3	105.0	73.0	79.0	83.6	85.8	86.6	83.5	0.1	2.3	3.1	
Mean	110.3	69.7	79.7	83.1	85.9	87.0	85.1	-2.0	0.8	1.9	ME
STD	6.1	2.9	3.1	1.4	1.6	1.9	1.6	1.9	1.6	1.5	STDE
								2.1	1.3	1.9	MAE

S05 D2	Omron Recordings			Omron Calculated MAP			MAP_{ratio}	Errors (Reference - MAP_{ratio})			
Rec #	SBP	DBP	HR	$MAP_{33\%}$	$MAP_{40\%}$	MAP_{HR}		$MAP_{33\%}$	$MAP_{40\%}$	MAP_{HR}	
1	118.0	70.0	87.0	85.8	89.2	90.9	92.4	-6.6	-3.2	-1.5	
2	119.0	71.0	86.0	86.8	90.2	91.8	94.1	-7.3	-3.9	-2.3	
3	119.0	70.0	85.0	86.2	89.6	91.2	89.4	-3.2	0.2	1.8	
Mean	118.7	70.3	86.0	86.3	89.7	91.3	92.0	-5.7	-2.3	-0.7	ME
STD	0.6	0.6	1.0	0.5	0.5	0.5	2.4	2.2	2.2	2.2	STDE
								5.7	2.4	1.9	MAE

S05 D3	Omron Recordings			Omron Calculated MAP			MAP_{ratio}	Errors (Reference - MAP_{ratio})			
Rec #	SBP	DBP	HR	$MAP_{33\%}$	$MAP_{40\%}$	MAP_{HR}		$MAP_{33\%}$	$MAP_{40\%}$	MAP_{HR}	
1	112.0	72.0	83.0	85.2	88.0	89.2	92.6	-7.4	-4.6	-3.4	
2	116.0	65.0	75.0	81.8	85.4	86.4	83.2	-1.4	2.2	3.2	
3	107.0	66.0	81.0	79.5	82.4	83.5	74.7	4.8	7.7	8.8	
Mean	111.7	67.7	79.7	82.2	85.3	86.4	83.5	-1.3	1.8	2.9	ME
STD	4.5	3.8	4.2	2.9	2.8	2.8	9.0	6.1	6.2	6.1	STDE
								4.5	4.8	5.2	MAE

S06 D1	Omron Recordings			Omron Calculated MAP			MAP_{ratio}	Errors (Reference - MAP_{ratio})			
Rec #	SBP	DBP	HR	$MAP_{33\%}$	$MAP_{40\%}$	MAP_{HR}		$MAP_{33\%}$	$MAP_{40\%}$	MAP_{HR}	
1	90.0	62.0	71.0	71.2	73.2	73.6	69.7	1.5	3.5	3.9	
2	96.0	66.0	66.0	75.9	78.0	78.3	72.0	3.9	6.0	6.3	
3	96.0	65.0	64.0	75.2	77.4	77.6	74.6	0.6	2.8	3.0	
Mean	94.0	64.3	67.0	74.1	76.2	76.5	72.1	2.0	4.1	4.4	ME
STD	3.5	2.1	3.6	2.5	2.6	2.5	2.5	1.7	1.7	1.7	STDE
								2.0	4.1	4.4	MAE

S06 D2	Omron Recordings			Omron Calculated MAP			MAP_{ratio}	Errors (Reference - MAP_{ratio})			
Rec #	SBP	DBP	HR	$MAP_{33\%}$	$MAP_{40\%}$	MAP_{HR}		$MAP_{33\%}$	$MAP_{40\%}$	MAP_{HR}	
1	91.0	62.0	70.0	71.6	73.6	74.0	69.3	2.3	4.3	4.7	
2	92.0	66.0	69.0	74.6	76.4	76.7	73.0	1.6	3.4	3.7	
3	98.0	66.0	63.0	76.6	78.8	79.0	73.9	2.7	4.9	5.1	
Mean	93.7	64.7	67.3	74.2	76.3	76.6	72.1	2.2	4.2	4.5	ME
STD	3.8	2.3	3.8	2.5	2.6	2.5	2.4	0.5	0.8	0.7	STDE
								2.2	4.2	4.5	MAE

S06 D3	Omron Recordings			Omron Calculated MAP			MAP_{ratio}	Errors (Reference - MAP_{ratio})			
Rec #	SBP	DBP	HR	$MAP_{33\%}$	$MAP_{40\%}$	MAP_{HR}		$MAP_{33\%}$	$MAP_{40\%}$	MAP_{HR}	
1	96.0	58.0	66.0	70.5	73.2	73.5	78.3	-7.8	-5.1	-4.8	
2	93.0	61.0	66.0	71.6	73.8	74.1	77.6	-6.0	-3.8	-3.5	
3	91.0	58.0	65.0	68.9	71.2	71.5	80.5	-11.6	-9.3	-9.0	
Mean	93.3	59.0	65.7	70.3	72.7	73.0	78.8	-8.5	-6.1	-5.8	ME
STD	2.5	1.7	0.6	1.3	1.4	1.4	1.5	2.9	2.9	2.9	STDE
								8.5	6.1	5.8	MAE

S07 D1	Omron Recordings			Omron Calculated MAP			MAP_{ratio}	Errors (Reference - MAP_{ratio})			
Rec #	SBP	DBP	HR	$MAP_{33\%}$	$MAP_{40\%}$	MAP_{HR}		$MAP_{33\%}$	$MAP_{40\%}$	MAP_{HR}	
1	115.0	77.0	61.0	89.5	92.2	92.3	92.0	-2.5	0.2	0.3	
2	112.0	76.0	54.0	87.9	90.4	90.2	92.2	-4.3	-1.8	-2.0	
3	113.0	75.0	52.0	87.5	90.2	89.9	88.6	-1.1	1.6	1.3	
Mean	113.3	76.0	55.7	88.3	90.9	90.8	90.9	-2.6	0.0	-0.1	ME
STD	1.5	1.0	4.7	1.1	1.1	1.3	2.0	1.6	1.7	1.7	STDE
								2.6	1.2	1.2	MAE

S07 D2	Omron Recordings			Omron Calculated MAP			MAP_{ratio}	Errors (Reference - MAP_{ratio})			
Rec #	SBP	DBP	HR	$MAP_{33\%}$	$MAP_{40\%}$	MAP_{HR}		$MAP_{33\%}$	$MAP_{40\%}$	MAP_{HR}	
1	105.0	76.0	51.0	85.6	87.6	87.3	89.5	-3.9	-1.9	-2.2	
2	106.0	78.0	44.0	87.2	89.2	88.7	92.2	-5.0	-3.0	-3.5	
3	107.0	73.0	46.0	84.2	86.6	86.1	92.1	-7.9	-5.5	-6.0	
Mean	106.0	75.7	47.0	85.7	87.8	87.4	91.3	-5.6	-3.5	-3.9	ME
STD	1.0	2.5	3.6	1.5	1.3	1.3	1.5	2.0	1.8	2.0	STDE
								5.6	3.5	3.9	MAE

S07 D3	Omron Recordings			Omron Calculated MAP			MAP_{ratio}	Errors (Reference - MAP_{ratio})			
Rec #	SBP	DBP	HR	$MAP_{33\%}$	$MAP_{40\%}$	MAP_{HR}		$MAP_{33\%}$	$MAP_{40\%}$	MAP_{HR}	
1	109.0	71.0	54.0	83.5	86.2	86.0	90.0	-6.5	-3.8	-4.0	
2	109.0	77.0	57.0	87.6	89.8	89.7	88.2	-0.6	1.6	1.5	
3	107.0	73.0	55.0	84.2	86.6	86.5	85.4	-1.2	1.2	1.1	
Mean	108.3	73.7	55.3	85.1	87.5	87.4	87.9	-2.8	-0.3	-0.5	ME
STD	1.2	3.1	1.5	2.2	2.0	2.0	2.3	3.2	3.0	3.1	STDE
								2.8	2.2	2.2	MAE

S08 D1	Omron Recordings			Omron Calculated MAP			MAP_{ratio}	Errors (Reference - MAP_{ratio})			
Rec #	SBP	DBP	HR	$MAP_{33\%}$	$MAP_{40\%}$	MAP_{HR}		$MAP_{33\%}$	$MAP_{40\%}$	MAP_{HR}	
1	121.0	74.0	66.0	89.5	92.8	93.2	92.6	-3.1	0.2	0.6	
2	126.0	74.0	67.0	91.2	94.8	95.3	92.4	-1.2	2.4	2.9	
3	126.0	74.0	72.0	91.2	94.8	95.7	96.1	-4.9	-1.3	-0.4	
Mean	124.3	74.0	68.3	90.6	94.1	94.7	93.7	-3.1	0.4	1.0	ME
STD	2.9	0.0	3.2	1.0	1.2	1.3	2.1	1.8	1.9	1.7	STDE
								3.1	1.3	1.3	MAE

S08 D2	Omron Recordings			Omron Calculated MAP			MAP_{ratio}	Errors (Reference - MAP_{ratio})			
Rec #	SBP	DBP	HR	$MAP_{33\%}$	$MAP_{40\%}$	MAP_{HR}		$MAP_{33\%}$	$MAP_{40\%}$	MAP_{HR}	
1	127.0	75.0	66.0	92.2	95.8	96.3	89.5	2.7	6.3	6.8	
2	127.0	71.0	66.0	89.5	93.4	93.9	88.5	1.0	4.9	5.4	
3	132.0	75.0	73.0	93.8	97.8	98.8	98.2	-4.4	-0.4	0.6	
Mean	128.7	73.7	68.3	91.8	95.7	96.3	92.1	-0.3	3.6	4.3	ME
STD	2.9	2.3	4.0	2.2	2.2	2.4	5.3	3.7	3.5	3.2	STDE
								2.7	3.9	4.3	MAE

S08 D3	Omron Recordings			Omron Calculated MAP			MAP_{ratio}	Errors (Reference - MAP_{ratio})			
Rec #	SBP	DBP	HR	$MAP_{33\%}$	$MAP_{40\%}$	MAP_{HR}		$MAP_{33\%}$	$MAP_{40\%}$	MAP_{HR}	
1	114.0	69.0	59.0	83.9	87.0	87.0	88.1	-4.3	-1.1	-1.1	
2	122.0	70.0	61.0	87.2	90.8	91.0	90.8	-3.6	0.0	0.2	
3	120.0	70.0	64.0	86.5	90.0	90.3	88.1	-1.6	1.9	2.2	
Mean	118.7	69.7	61.3	85.8	89.3	89.4	89.0	-3.2	0.3	0.4	ME
STD	4.2	0.6	2.5	1.8	2.0	2.1	1.6	1.4	1.5	1.7	STDE
								3.2	1.0	1.2	MAE

S09 D1	Omron Recordings			Omron Calculated MAP			MAP_{ratio}	Errors (Reference - MAP_{ratio})			
Rec #	SBP	DBP	HR	$MAP_{33\%}$	$MAP_{40\%}$	MAP_{HR}		$MAP_{33\%}$	$MAP_{40\%}$	MAP_{HR}	
1	100.0	73.0	62.0	81.9	83.8	83.9	83.7	-1.8	0.1	0.2	
2	98.0	71.0	66.0	79.9	81.8	82.0	84.9	-5.0	-3.1	-2.9	
3	110.0	79.0	64.0	89.2	91.4	91.6	93.9	-4.7	-2.5	-2.3	
Mean	102.7	74.3	64.0	83.7	85.7	85.9	87.5	-3.8	-1.8	-1.6	ME
STD	6.4	4.2	2.0	4.9	5.1	5.1	5.6	1.8	1.7	1.6	STDE
								3.8	1.9	1.8	MAE

S09 D2	Omron Recordings			Omron Calculated MAP			MAP_{ratio}	Errors (Reference - MAP_{ratio})			
Rec #	SBP	DBP	HR	$MAP_{33\%}$	$MAP_{40\%}$	MAP_{HR}		$MAP_{33\%}$	$MAP_{40\%}$	MAP_{HR}	
1	108.0	73.0	64.0	84.6	87.0	87.2	89.0	-4.5	-2.0	-1.8	
2	102.0	75.0	70.0	83.9	85.8	86.2	88.4	-4.5	-2.6	-2.2	
3	102.0	71.0	65.0	81.2	83.4	83.6	81.9	-0.7	1.5	1.7	
Mean	104.0	73.0	66.3	83.2	85.4	85.7	86.4	-3.2	-1.0	-0.7	ME
STD	3.5	2.0	3.2	1.8	1.8	1.8	3.9	2.2	2.2	2.2	STDE
								3.2	2.0	1.9	MAE

S09 D3	Omron Recordings			Omron Calculated MAP			MAP_{ratio}	Errors (Reference - MAP_{ratio})			
Rec #	SBP	DBP	HR	$MAP_{33\%}$	$MAP_{40\%}$	MAP_{HR}		$MAP_{33\%}$	$MAP_{40\%}$	MAP_{HR}	
1	96.0	67.0	69.0	76.6	78.6	79.0	81.8	-5.2	-3.2	-2.8	
2	102.0	68.0	64.0	79.2	81.6	81.8	79.7	-0.5	1.9	2.1	
3	96.0	71.0	69.0	79.3	81.0	81.3	80.7	-1.5	0.3	0.6	
Mean	98.0	68.7	67.3	78.3	80.4	80.7	80.7	-2.4	-0.3	0.0	ME
STD	3.5	2.1	2.9	1.5	1.6	1.5	1.1	2.5	2.6	2.5	STDE
								2.4	1.8	1.9	MAE

S10 D1	Omron Recordings			Omron Calculated MAP			MAP_{ratio}	Errors (Reference - MAP_{ratio})			
Rec #	SBP	DBP	HR	$MAP_{33\%}$	$MAP_{40\%}$	MAP_{HR}		$MAP_{33\%}$	$MAP_{40\%}$	MAP_{HR}	
1	93.0	67.0	76.0	75.6	77.4	78.0	98.5	-22.9	-21.1	-20.5	
2	92.0	66.0	71.0	74.6	76.4	76.8	97.8	-23.2	-21.4	-21.0	
3	91.0	62.0	69.0	71.6	73.6	74.0	93.8	-22.2	-20.2	-19.8	
Mean	92.0	65.0	72.0	73.9	75.8	76.2	96.7	-22.8	-20.9	-20.5	ME
STD	1.0	2.6	3.6	2.1	2.0	2.0	2.5	0.5	0.6	0.6	STDE
								22.8	20.9	20.5	MAE

S10 D2	Omron Recordings			Omron Calculated MAP			MAP_{ratio}	Errors (Reference - MAP_{ratio})			
Rec #	SBP	DBP	HR	$MAP_{33\%}$	$MAP_{40\%}$	MAP_{HR}		$MAP_{33\%}$	$MAP_{40\%}$	MAP_{HR}	
1	91.0	59.0	76.0	69.6	71.8	72.5	97.7	-28.1	-25.9	-25.2	
2	91.0	60.0	82.0	70.2	72.4	73.3	98.5	-28.3	-26.1	-25.2	
3	92.0	58.0	78.0	69.2	71.6	72.4	97.9	-28.7	-26.3	-25.5	
Mean	91.3	59.0	78.7	69.7	71.9	72.7	98.0	-28.4	-26.1	-25.3	ME
STD	0.6	1.0	3.1	0.5	0.4	0.5	0.4	0.3	0.2	0.2	STDE
								28.4	26.1	25.3	MAE

S10 D3	Omron Recordings			Omron Calculated MAP			MAP_{ratio}	Errors (Reference - MAP_{ratio})			
Rec #	SBP	DBP	HR	$MAP_{33\%}$	$MAP_{40\%}$	MAP_{HR}		$MAP_{33\%}$	$MAP_{40\%}$	MAP_{HR}	
1	102.0	61.0	91.0	74.5	77.4	79.0	100.0	-25.5	-22.6	-21.0	
2	100.0	63.0	86.0	75.2	77.8	79.0	98.7	-23.5	-20.9	-19.7	
3	101.0	61.0	87.0	74.2	77.0	78.4	99.2	-25.0	-22.2	-20.8	
Mean	101.0	61.7	88.0	74.6	77.4	78.8	99.3	-24.7	-21.9	-20.5	ME
STD	1.0	1.2	2.6	0.5	0.4	0.4	0.7	1.0	0.9	0.7	STDE
								24.7	21.9	20.5	MAE

Appendix E - Experimental Results with Dataset of Patients

Each table block refers to one patient. P01 refers to patient #1. The first column shows the patient ID and recording number. The next three columns show the values displayed by the BpTRU monitor and the next three columns show the three MAP values calculated from them. Next column shows MAP estimated using the proposed approach and the last three columns show the difference between reference MAP and MAP estimated using the proposed approach.

P01	BpTRU Recordings			BpTRU Calculated MAP			MAP_{ratio}	Errors (Reference - MAP_{ratio})			
Rec #	SBP	DBP	HR	$MAP_{33\%}$	$MAP_{40\%}$	MAP_{HR}		$MAP_{33\%}$	$MAP_{40\%}$	MAP_{HR}	
1	114.0	70.0	75.0	84.5	87.6	88.5	79.8	4.7	7.8	8.7	
2	108.0	69.0	67.0	81.9	84.6	85.0	82.8	-0.9	1.8	2.2	
3	113.0	57.0	67.0	75.5	79.4	80.0	81.8	-6.3	-2.4	-1.8	
4	112.0	51.0	72.0	71.1	75.4	76.4	77.7	-6.6	-2.3	-1.3	
Mean	111.8	61.8	70.3	78.3	81.8	82.5	80.5	-2.3	1.2	1.9	ME
STD	2.6	9.3	3.9	6.1	5.4	5.3	2.3	5.3	4.8	4.8	STDE
								4.6	3.6	3.5	MAE

P02	BpTRU Recordings			BpTRU Calculated MAP			MAP_{ratio}	Errors (Reference - MAP_{ratio})			
Rec #	SBP	DBP	HR	$MAP_{33\%}$	$MAP_{40\%}$	MAP_{HR}		$MAP_{33\%}$	$MAP_{40\%}$	MAP_{HR}	
1	103.0	83.0	84.0	89.6	91.0	91.6	84.9	4.7	6.1	6.7	
2	106.0	80.0	109.0	88.6	90.4	92.0	85.8	2.8	4.6	6.2	
3	99.0	75.0	88.0	82.9	84.6	85.5	83.2	-0.3	1.4	2.3	
4	110.0	75.0	75.0	86.6	89.0	89.7	87.1	-0.5	1.9	2.6	
Mean	104.5	78.3	89.0	86.9	88.8	89.7	85.3	1.7	3.5	4.4	ME
STD	4.7	3.9	14.4	2.9	2.9	3.0	1.6	2.5	2.2	2.3	STDE
								2.1	3.5	4.4	MAE

P03	BpTRU Recordings			BpTRU Calculated MAP			MAP_{ratio}	Errors (Reference - MAP_{ratio})			
Rec #	SBP	DBP	HR	$MAP_{33\%}$	$MAP_{40\%}$	MAP_{HR}		$MAP_{33\%}$	$MAP_{40\%}$	MAP_{HR}	
1	159.0	90.0	75.0	112.8	117.6	119.0	108.8	4.0	8.8	10.2	
2	143.0	85.0	75.0	104.1	108.2	109.4	102.9	1.2	5.3	6.5	
3	131.0	81.0	75.0	97.5	101.0	102.0	104.5	-7.0	-3.5	-2.5	
4	128.0	83.0	74.0	97.9	101.0	101.8	105.4	-7.6	-4.4	-3.6	
Mean	140.3	84.8	74.8	103.1	107.0	108.0	105.4	-2.3	1.6	2.6	ME
STD	14.1	3.9	0.5	7.2	7.9	8.1	2.5	5.8	6.5	6.7	STDE
								4.9	5.5	5.7	MAE

P04	BpTRU Recordings			BpTRU Calculated MAP			MAP_{ratio}	Errors (Reference - MAP_{ratio})			
Rec #	SBP	DBP	HR	$MAP_{33\%}$	$MAP_{40\%}$	MAP_{HR}		$MAP_{33\%}$	$MAP_{40\%}$	MAP_{HR}	
1	139.0	67.0	64.0	90.8	95.8	96.3	86.7	4.1	9.1	9.6	
2	137.0	66.0	64.0	89.4	94.4	94.9	82.8	6.6	11.6	12.1	
3	123.0	56.0	63.0	78.1	82.8	83.2	84.9	-6.8	-2.1	-1.7	
4	127.0	61.0	64.0	82.8	87.4	87.8	82.7	0.1	4.7	5.1	
Mean	131.5	62.5	63.8	85.3	90.1	90.5	84.3	1.0	5.8	6.3	ME
STD	7.7	5.1	0.5	5.9	6.1	6.1	1.9	5.8	6.0	6.1	STDE
								4.4	6.9	7.1	MAE

P05	BpTRU Recordings			BpTRU Calculated MAP			MAP_{ratio}	Errors (Reference - MAP_{ratio})			
Rec #	SBP	DBP	HR	$MAP_{33\%}$	$MAP_{40\%}$	MAP_{HR}		$MAP_{33\%}$	$MAP_{40\%}$	MAP_{HR}	
1	150.0	84.0	70.0	105.8	110.4	111.3	89.9	15.9	20.5	21.4	
2	151.0	78.0	68.0	102.1	107.2	108.0	88.7	13.4	18.5	19.3	
3	136.0	81.0	69.0	99.2	103.0	103.7	90.0	9.2	13.0	13.7	
4	145.0	81.0	67.0	102.1	106.6	107.3	88.8	13.3	17.8	18.5	
Mean	145.5	81.0	68.5	102.3	106.8	107.6	89.4	12.9	17.5	18.2	ME
STD	6.9	2.4	1.3	2.7	3.0	3.1	0.7	2.8	3.2	3.3	STDE
								12.9	17.5	18.2	MAE

P07	BpTRU Recordings			BpTRU Calculated MAP			MAP_{ratio}	Errors (Reference - MAP_{ratio})			
Rec #	SBP	DBP	HR	$MAP_{33\%}$	$MAP_{40\%}$	MAP_{HR}		$MAP_{33\%}$	$MAP_{40\%}$	MAP_{HR}	
1	111.0	65.0	64.0	80.2	83.4	83.7	85.6	-5.4	-2.2	-1.9	
2	113.0	59.0	65.0	76.8	80.6	81.0	82.4	-5.6	-1.8	-1.4	
3	110.0	67.0	61.0	81.2	84.2	84.3	85.8	-4.6	-1.6	-1.5	
4	122.0	67.0	63.0	85.2	89.0	89.3	82.2	3.0	6.8	7.1	
Mean	114.0	64.5	63.3	80.8	84.3	84.6	84.0	-3.2	0.3	0.6	ME
STD	5.5	3.8	1.7	3.4	3.5	3.5	2.0	4.1	4.3	4.3	STDE
								4.6	3.1	3.0	MAE

P08	BpTRU Recordings			BpTRU Calculated MAP			MAP_{ratio}	Errors (Reference - MAP_{ratio})			
Rec #	SBP	DBP	HR	$MAP_{33\%}$	$MAP_{40\%}$	MAP_{HR}		$MAP_{33\%}$	$MAP_{40\%}$	MAP_{HR}	
1	124.0	93.0	72.0	103.2	105.4	105.9	107.0	-3.8	-1.6	-1.1	
2	139.0	87.0	72.0	104.2	107.8	108.7	112.4	-8.2	-4.6	-3.7	
3	126.0	93.0	75.0	103.9	106.2	106.9	106.7	-2.8	-0.5	0.2	
4	148.0	102.0	67.0	117.2	120.4	120.9	112.8	4.4	7.6	8.1	
Mean	134.3	93.8	71.5	107.1	110.0	110.6	109.7	-2.6	0.2	0.8	ME
STD	11.3	6.2	3.3	6.7	7.0	7.0	3.3	5.2	5.2	5.1	STDE
								4.8	3.6	3.3	MAE

P09	BpTRU Recordings			BpTRU Calculated MAP			MAP_{ratio}	Errors (Reference - MAP_{ratio})			
Rec #	SBP	DBP	HR	$MAP_{33\%}$	$MAP_{40\%}$	MAP_{HR}		$MAP_{33\%}$	$MAP_{40\%}$	MAP_{HR}	
1	112.0	66.0	68.0	81.2	84.4	84.9	83.5	-2.3	0.9	1.4	
2	120.0	66.0	67.0	83.8	87.6	88.2	83.3	0.5	4.3	4.9	
3	109.0	62.0	68.0	77.5	80.8	81.3	77.1	0.4	3.7	4.2	
4	104.0	63.0	67.0	76.5	79.4	79.8	81.4	-4.9	-2.0	-1.6	
Mean	111.3	64.3	67.5	79.8	83.1	83.6	81.3	-1.6	1.7	2.2	ME
STD	6.7	2.1	0.6	3.4	3.7	3.7	3.0	2.6	2.9	2.9	STDE
								2.0	2.7	3.0	MAE

P10	BpTRU Recordings			BpTRU Calculated MAP			MAP_{ratio}	Errors (Reference - MAP_{ratio})			
Rec #	SBP	DBP	HR	$MAP_{33\%}$	$MAP_{40\%}$	MAP_{HR}		$MAP_{33\%}$	$MAP_{40\%}$	MAP_{HR}	
1	125.0	90.0	74.0	101.6	104.0	104.7	99.5	2.1	4.5	5.2	
2	133.0	83.0	74.0	99.5	103.0	103.9	97.9	1.6	5.1	6.0	
3	126.0	89.0	69.0	101.2	103.8	104.3	99.3	1.9	4.5	5.0	
4	129.0	75.0	70.0	92.8	96.6	97.4	102.5	-9.7	-5.9	-5.1	
Mean	128.3	84.3	71.8	98.8	101.9	102.6	99.8	-1.0	2.1	2.8	ME
STD	3.6	6.9	2.6	4.1	3.5	3.5	1.9	5.8	5.3	5.3	STDE
								3.8	5.0	5.3	MAE

P11	BpTRU Recordings			BpTRU Calculated MAP			MAP_{ratio}	Errors (Reference - MAP_{ratio})			
Rec #	SBP	DBP	HR	$MAP_{33\%}$	$MAP_{40\%}$	MAP_{HR}		$MAP_{33\%}$	$MAP_{40\%}$	MAP_{HR}	
1	132.0	95.0	76.0	107.2	109.8	110.6	111.5	-4.3	-1.7	-0.9	
2	135.0	86.0	76.0	102.2	105.6	106.6	107.7	-5.5	-2.1	-1.1	
3	142.0	99.0	75.0	113.2	116.2	117.1	107.5	5.7	8.7	9.6	
4	149.0	97.0	75.0	114.2	117.8	118.8	111.1	3.1	6.7	7.7	
Mean	139.5	94.3	75.5	109.2	112.4	113.3	109.5	-0.3	2.9	3.8	ME
STD	7.6	5.7	0.6	5.6	5.7	5.7	2.1	5.5	5.6	5.6	STDE
								4.6	4.8	4.8	MAE

P12	BpTRU Recordings			BpTRU Calculated MAP			MAP_{ratio}	Errors (Reference - MAP_{ratio})			
Rec #	SBP	DBP	HR	$MAP_{33\%}$	$MAP_{40\%}$	MAP_{HR}		$MAP_{33\%}$	$MAP_{40\%}$	MAP_{HR}	
1	139.0	94.0	96.0	108.9	112.0	114.0	106.3	2.6	5.7	7.7	
2	123.0	84.0	97.0	96.9	99.6	101.4	107.2	-10.3	-7.6	-5.8	
3	135.0	93.0	90.0	106.9	109.8	111.4	100.6	6.3	9.2	10.8	
4	138.0	84.0	87.0	101.8	105.6	107.5	98.2	3.6	7.4	9.3	
Mean	133.8	88.8	92.5	103.6	106.8	108.6	103.1	0.5	3.7	5.5	ME
STD	7.4	5.5	4.8	5.4	5.5	5.5	4.4	7.4	7.7	7.6	STDE
								5.7	7.5	8.4	MAE

P13	BpTRU Recordings			BpTRU Calculated MAP				Errors (Reference - MAP_{ratio})			
Rec #	SBP	DBP	HR	$MAP_{33\%}$	$MAP_{40\%}$	MAP_{HR}	MAP_{ratio}	$MAP_{33\%}$	$MAP_{40\%}$	MAP_{HR}	
1	120.0	65.0	51.0	83.2	87.0	86.5	85.8	-2.6	1.2	0.7	
2	108.0	62.0	59.0	77.2	80.4	80.4	86.4	-9.2	-6.0	-6.0	
3	125.0	62.0	61.0	82.8	87.2	87.4	88.8	-6.0	-1.6	-1.4	
4	122.0	67.0	53.0	85.2	89.0	88.6	92.5	-7.3	-3.5	-3.9	
Mean	118.8	64.0	56.0	82.1	85.9	85.8	88.4	-6.3	-2.5	-2.6	ME
STD	7.5	2.4	4.8	3.4	3.8	3.6	3.0	2.8	3.0	2.9	STDE
								6.3	3.1	3.0	MAE

P14	BpTRU Recordings			BpTRU Calculated MAP				Errors (Reference - MAP_{ratio})			
Rec #	SBP	DBP	HR	$MAP_{33\%}$	$MAP_{40\%}$	MAP_{HR}	MAP_{ratio}	$MAP_{33\%}$	$MAP_{40\%}$	MAP_{HR}	
1	93.0	70.0	102.0	77.6	79.2	80.4	77.5	0.1	1.7	2.9	
2	108.0	69.0	101.0	81.9	84.6	86.6	73.0	8.9	11.6	13.6	
3	101.0	61.0	102.0	74.2	77.0	79.1	72.0	2.2	5.0	7.1	
4	100.0	61.0	101.0	73.9	76.6	78.6	71.6	2.3	5.0	7.0	
Mean	100.5	65.3	101.5	76.9	79.4	81.2	73.5	3.4	5.8	7.6	ME
STD	6.1	4.9	0.6	3.7	3.7	3.7	2.7	3.8	4.2	4.4	STDE
								3.4	5.8	7.6	MAE

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