NONINVASIVE BLOOD PRESSURE MEASUREMENT METHODS

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This paper review the historical background of noninvasive blood pressure measurement methods with their accuracies achieved and explains the requirements for common national and international standards of accuracy. Also are shortly described the development of blood pressure measurement system and a method, named electronic palpation method (EP), suitable for the long-term measurement of heart rate and blood pressure using a cuff on the upper arm and a pressure sensor array on the radial artery.

Keywords: pressure transducer, blood pressure monitoring, pulse transit time

1. INTRODUCTION

Cardiovascular diseases are one of the leading causes of death. Globally, they underlie the death of one third of the world’s population and among Europeans, such as the Finnish, one half of all deaths are attributed to them. Noninvasive measurements for assessing these diseases would be highly valuable for obvious reasons.

Invasive blood pressure measurement methods are generally used in hospitals, especially in intensive care units, are known to carry a risk, albeit a small one. Risks of these kinds could be avoided, however, if there was a noninvasive method offering a high degree of accuracy and real time operation in a continuous, beat-to-beat, mode. Further, the method should be insensitive to the patient’s movement (artifacts) and respond rapidly to cardiovascular changes, such as a sudden drop in blood pressure. As it is, invasive methods are currently prevalent.

In clinics and home care, noninvasive blood pressure measurement devices have become increasingly common during the last decade as their prices have sunk to an appropriate level for ordinary consumers. Automatic measurement features and easiness of use have also contributed
to their growing popularity. Nevertheless, the accuracy of these devices has not yet reached the necessary level, since only some of them are clinically validated and most have a questionable accuracy [1].

2. STANDARDS AND ACCURACY VALIDATION

In 1987, the American Association for the Advancement of Medical Instrumentation, AAMI, published a standard for sphygmomanometers, which included a protocol for evaluating the accuracy of devices [2, 3]. In 1990, a protocol was devised by the British Hypertension Society, BHS, [4, 5] and later by the European Society of Hypertension, ESH [6]. This was then used as the basis for the International Protocol [7].

According to the AAMI standard, the mean difference between different measurements must be less than 5 mmHg, or the standard deviation must be less than 8 mmHg with 85% of the measurements and in the 20-250 mmHg range. Accuracy better than 10 mmHg must be achieved with 95% of the measurements.

In the British protocol, accuracy falls into four grades, A to D, where A denotes the greatest and D the least agreement. These grades represent the cumulative percentage of readings falling within 5, 10 and 15 mmHg. To fulfil the BHS protocol, a device must achieve at least grade B. As the table below indicates, grades in the British protocol are less strict than in the AAMI standard. Grades are specified separately for diastolic and systolic blood pressure.

<table>
<thead>
<tr>
<th>Grade</th>
<th>≤5</th>
<th>≤10</th>
<th>≤15</th>
</tr>
</thead>
<tbody>
<tr>
<td>A</td>
<td>60%</td>
<td>85%</td>
<td>95%</td>
</tr>
<tr>
<td>B</td>
<td>50%</td>
<td>75%</td>
<td>90%</td>
</tr>
<tr>
<td>C</td>
<td>40%</td>
<td>65%</td>
<td>85%</td>
</tr>
<tr>
<td>D</td>
<td></td>
<td></td>
<td>Worse than C</td>
</tr>
</tbody>
</table>

Put together by a working group of the European Society of Hypertension, the International Protocol for Validation of Blood Pressure Measuring Devices in Adults provides a recommendation for the validation of such devices. Using a standard mercury sphygmomanometer as a reference, the protocol requires that 33 study subjects, 15 in Phase 1 and 15 plus 18 in Phase 2. Seven measurements must be performed, starting from measurements carried out by two observers (reference), which will be averaged, followed by a supervisor’s test (device under test). Four reference measurement pairs and three supervisor’s measurements must be conducted on each patient in turn. By subtracting the mean values of the obtained reference pair measurements from the supervisor’s value, we will arrive at the difference. All subjects should be at least 30 years of age and may be taking antihypertensive medication. These accuracy criteria are quite similar to those of the BHS: categories for
rounded values are ≤ 5 mmHg (no error of clinical relevance), 6-10 mmHg (slightly inaccurate), 11–15 mmHg (moderately inaccurate) and ≥ 15 mmHg (very inaccurate). The table below can be used for validation. For clinical approval, the device under test must pass all levels.

Table 2. Grading criteria used by the International Protocol. [7]

<table>
<thead>
<tr>
<th>Grade</th>
<th>Rounded absolute difference between standard and test device</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>within 5 mmHg</td>
</tr>
<tr>
<td>Phase 1</td>
<td>At least one of three</td>
</tr>
<tr>
<td>(15x3 = 45 pairs)</td>
<td>25</td>
</tr>
<tr>
<td>Phase 2.1</td>
<td>Two of three</td>
</tr>
<tr>
<td>(33x3 = 99 pairs)</td>
<td>65</td>
</tr>
<tr>
<td></td>
<td>All of</td>
</tr>
<tr>
<td></td>
<td>60</td>
</tr>
<tr>
<td>Phase 2.2</td>
<td>2/3 within 5 mmHg</td>
</tr>
</tbody>
</table>

3. NONINVASIVE BLOOD PRESSURE MEASUREMENT METHODS

A historical overview of the different noninvasive blood pressure measurement methods will be given here together with a description of each. Most of the historical data are gleaned from Geddes’ Handbook of Blood Pressure Measurement, published in 1991 [8], complemented by new information put out after its publication. Particular emphasis will be placed on detailing the methods’ strong and weak points and on the achieved accuracies. Towards the end, the accuracy of a number of blood pressure measurement devices will be presented. The pressure unit used systematically in this study is mmHg, which, although not an SI unit, is nonetheless a globally approved blood pressure unit for historical reasons.

Counterpressure with sphygmomanometer

Noninvasive blood pressure measurement methods are indirect and based mainly on measuring counterpressure. Putting a hand in a water-filled chamber, and sealing it with the wrist, allows the measurement of pressure alterations in the chamber. This pressure alteration, caused by blood pressure oscillations in the hand, depends on the static pressure within the chamber. If this pressure is increased, the pressure alteration will rise, reach its maximum, and then decrease until blood circulation in the hand stops altogether. In 1876, Etienne Jules Marey, the French physiologist, first used this method for noninvasive blood pressure assessment in humans.

In the same year, Von Bash employed a water-filled bag that was connected to a mercury manometer. When this bag was pressed with increasing force against the skin over the radial artery, pulsations could be observed until, after a certain point, they disappeared. At this point, counterpressure was assumed to represent systolic pressure.
Hydraulic counterpressure was subsequently replaced by pneumatic pressure. Marey designed an instrument, consisting of a rubber sleeve in a rigid cylinder, a mercury manometer and a pump. Once a finger was attached into the cylinder, weak pulsations could be observed in the mercury level. Potain improved this measurement system in 1902, by replacing the mercury manometer with a gauge and substituting water for air.

In 1895, Mosso improved Marey’s measurement system by applying counterpressure on all fingers, thereby obtaining a larger signal.

Almost at the same time, Scipione Riva Rocci (1896) in Italy and Hill and Barnard (1897) in England developed a cuff, which was attached around the upper arm, thus occluding the brachial artery when pressurized. In both these methods, the used cuffs were too narrow according to Von Recklinghausen (1901).

In 1899, Gärtner introduced a method to determine systolic blood pressure by placing a cuff around a finger, pressing blood off from the distal side of the cuff, inflating it to a point above presumed systolic pressure, and then deflating cuff pressure. When cuff pressure reached the systolic blood pressure level, blood flushed to the finger tip. Gärtner’s method has since been known as the blush or flush method.

As the pneumatic sleeve or cuff, also referred to as a sphygmomanometer, consisted usually of a rubber bladder woven inside a fabric, it enabled indirect, noninvasive arterial blood pressure measurements. Up until the present day, the cuff has been the single most important component of noninvasive blood pressure measurement devices. The cuff is usually attached around the arm on the brachial artery or, in some devices, around the wrist or even a single finger. Pressure measurements can be made by using a mercury or aneroid manometer, which have become the “gold standards” in clinical use, but are now being replaced with silicon-based pressure sensors inside automatic measurement devices, based usually on the auscultatory or oscillometric method.
Auscultatory method

In his MD thesis of 1905, Nikolai Korotkoff, from St. Petersburg, Russia, presented blood pressure measurements performed on animals using a Riva-Rocci sleeve, mercury manometer and child stethoscope. First, he quickly raised cuff pressure until it stopped blood circulation on the distal side of the hand, indicated by palpating the radial artery. During the following slow pressure drop, audible sounds could be heard through the stethoscope, which was placed on the skin beyond the sleeve. These sounds were affected by the blood wave in the artery under the cuff, and were audible at 10-12 mmHg, slightly before the pulse could be palpated on the radial artery. At this point, cuff pressure is taken to indicate maximum blood pressure, while minimum blood pressure is achieved when the murmur sounds disappear. (Geddes 1991)

Korotkoff sounds were corroborated by the British researchers MacWilliam and Melvin (1914) and the American Warfield (1912), who used intra-arterial pressure measured from a dog as a reference. Then in 1932, Wold and von Bonsdorff applied Korotkoff’s auscultatory method on humans using intra-arterial blood pressure as a reference.

Later, the characteristic sounds heard during cuff deflation were divided into five phases on the basis of their intensity. First to publish were Goodman and Howell (1911), followed by Grödel and Miller (1943), Korns (1926) as well as Rappaport and Luisada (1944).

The appearance and disappearance of sound can be used to determine systolic and diastolic blood pressure, respectively, while the cuff deflates. However, establishing the point at which sounds disappear is not always obvious; in fact, misleading readings are easy to record. Thus, a certain sound intensity level is often used to determine the point corresponding to diastolic blood pressure. Also the sound amplitude maximum can be used to determine mean arterial blood pressure, as demonstrated by Davis and Geddes (1989 and 1990).

The origin of the sounds has been discussed since Korotkoff’s initial findings. McCutcheon and Rushmer have provided the latest universally acknowledged explanation (1967): “An acceleration transient produced by an abrupt distension in the arterial wall as a jet of blood surges under the cuff into the distal side of the artery. This produces the first tapping sound which signals systolic pressure. It continues as cuff pressure is decreased and disappears at diastolic pressure.” Also, they mentioned that “Turbulent or eddy flow, which follows the initial jet, and produces audible sound. This factor plays little or no significance in the auscultatory technique.”

It should be pointed out that the genesis of the sound will be different in the four phases; one mechanism predominating in one phase and a different one in another. Geddes (1991) summarised in his Handbook of Blood Pressure Measurement that other factors can also be pointed out, such as the rate of the pressure increase, which has a direct effect on sound
intensity (Ehret 1909, Bramwell 1926 and Tavel 1969). As a result, the auscultatory method may yield erroneous results in hypertension (Rodbard 1962, 1967 Pederson and Vogt 1973). In addition, fluid velocity plays an important role, as explained by Flack (1915). Thus, a sufficient flow velocity is a precondition for the audibility of Korotkoff sounds (Rappaport 1944, Rodbard 1953, and McCutcheon 1967).

Measurements based on the auscultatory method are difficult to automate, because the frequency spectrum of the different phases of Korotkoff sounds is closely related to blood pressure. When a patient’s blood pressure is high, also the recorded frequency spectrum is higher than normal and decreases as a function of blood pressure. With hypotensive patients and infants, on the other hand, the highest spectrum components can be as low as 8 Hz (Whitcher et al. 1966 and 1967), which is below the human hearing bandwidth. Normotensive subjects, in turn, require a bandwidth of 20 Hz to 300 Hz for a sufficient reproduction of Korotkoff sounds (Geddes 1991), although most of the energy of the signal spectrum is below 100 Hz.

Since technology improvements have enabled efficient signal processing and calculation, a number of papers have been published on the classification of Korotkoff sounds. Cozby and Adhami [9], for example, discovered in 1993 the sub-audible, low frequency part of Korotkoff sounds and concluded that energy in the 1-10 Hz bandwidth of the total energy rises from 60% to 90%, when cuff pressure decreases from above the systolic to a level below it. This feature can be used as a thresholding algorithm for determining systolic blood pressure. In 1998 Rogueiro-Gómez and Pallás-Areny [10] were able to use the spectral energy dispersion ratio to determine systolic and diastolic blood pressure with high precision: in 97% of all cases, the determined values for 15 persons were within ±1 heart beat.

In ambulatory measurements, when the patient is able to move moderately, noise may become dominant, thereby spoiling the measurement. This may be avoided by using two identical microphones under the cuff, one located on the upper side, the other on the distal side. Ambient noise reaches both microphones at the same time, but the blood pressure pulse propagating through the brachial artery arrives after a time delay. This phenomenon can be used for noise cancellation as described by Sebald et al (2002) [11].
Oscillometric method

As discussed previously, the oscillometric method was first introduced by Marey in 1876. Erlanger (1904) further developed it and attached a Riva-Rocci cuff around the upper arm instead of fingers. Pressure oscillations were recorded on a rotating drum. This graphic recording was discarded by Pachon (1909), who used dual-dial gauges, one for presenting oscillation amplitude and the other for showing cuff pressure.

In those days, it was thought that maximum oscillation amplitude indicated diastolic blood pressure (Howell and Brush 1901). It was more than half a century later that Posey and Geddes showed that that point actually corresponds to mean arterial pressure. Ramsey [12] (1979) and Yelderman and Ream [13] (1979) verified the finding with adults, while Kimble et al [14] did the same with newborn infants in 1981. At around the same time, Alexander et al [15] demonstrated that the cuff width should be at least 40% of the arm circumference.

Diastolic and systolic blood pressures can be determined using special fractions of the maximum oscillation amplitude, also known as characteristic ratios. Thus, the special fractions for systolic and diastolic blood pressure are 50% and 80%, respectively. Friesen and Lichter [16] in 1981 used these characteristic ratios to determine the diastolic and systolic blood pressures of infants, neonatal and pre-term babies with excellent results: the regression line formula was $D = 0.94p + 3.53$ for systolic pressure and $D = 0.98p + 1.7$ for diastolic pressure.

Geddes et al [17] (1983) made measurements with animals (13 dogs) using the direct invasive method as a reference and with humans (43 adult subjects) using the auscultatory method as a reference. Auscultatory signals inside the cuff were recorded using a tiny piezoelectric microphone with a bandwidth of 30-300 Hz to record Korotkoff sounds and another with a bandwidth of 0.3-30 Hz to obtain cuff oscillations, both recorded simultaneously. For humans, the characteristic ratios were 45-57% and 75-86% for systolic and diastolic pressure, respectively. For animals, these values varied between 43-73% and 69-83%. Inevitably, these ratios vary considerably and more theoretical studies are needed.

Drzewiecki et al [18] (1994) used a mathematical model to study the theory that the oscillation maximum appears at mean arterial pressure. Their model supported the hypothesis. In addition, they found the average characteristic detection ratios to be 59% for systolic and 72% for diastolic pressure. They also suggest that the systolic ratio should be lower for hypertensive patients and that the diastolic ratio should be lower for hypotensive patients.

Ursino & Cristalli also conducted experimental studies using a pneumatic system. Later, they constructed mathematical models to describe pressure distribution and the influence it has on tissue properties under the cuff. They showed that the rigidity of the arterial wall and tissue compliance have a significant effect on the accuracy of diastolic and systolic pressure values,
causing an error of 15%-20%. By contrast, changes in mean arterial pressure and cuff compliance did not influence the measurement to a large degree. They also supposed that the mean arterial blood pressure can be determined as the lowest pressure at which cuff pulse amplitude reaches the plateau. Another finding by Ursino and Cristalli was that the characteristic ratios are also greatly affected by the arterial wall’s elasticity; thus, excessively elastic arteries may lower the systolic ratio by 25%-30% and this figure may rise to 80% with stiffened arteries. According to their model, arterial stiffness has a considerably weaker effect on the diastolic ratio. Typical characteristic ratio values were 46%-64% for the systolic ratio and 59%-82% for the diastolic ratio. As a conclusion, a measurement device with fixed ratios for determining systolic, mean and diastolic blood pressure may significantly overestimate them. [19]

Moraes and Cerulli (1999 and 2000) also studied the characteristic ratios using computer-controlled linear cuff pressure deflation with 10 patients and 75 volunteers, using the auscultatory method as a reference [20, 21]. A fixed percentile rule yielded a value of 56% for systolic and 76% for diastolic pressure. With these ratios, they got an average errors and standard deviations of error of (-0.9 ± 7.0) mmHg and (1.0 ± 6.5) mmHg for systolic and diastolic blood pressure, respectively. This format (average ± standard deviation) will be used anytime in the text if not specified in other terms. Applying an adaptive classification rule, they changed the ratios in conformity with the arm circumference and mean blood pressure, i.e., the systolic ratio was decreased from 64% to 29%, mean artery pressure was changed from less than 70 mmHg to over 150 mmHg and diastolic pressure was increased from 50% to 75%. The obtained accuracies were (-1.5 ± 5.1) mmHg for systolic and (0.6 ± 5.9) mmHg for diastolic pressure. As can be seen, the standard deviation decreased slightly. In general, the deviations in the characteristic ratios recorded by Moraes and Cerulli ranged widely.

Being dependent on the used ratios, the fixed percentile rule will lead to erroneous readings in many measurements. In addition, motion artefacts, the white coat effect, shivering and arrhythmia generate noise that has an adverse effect on the accuracy of the readings. Because most manufacturers of automatic devices do not reveal exactly how their device determines the pressure values, the only option available to researchers is to test them with a large number of patients with different conditions, including hypo-, normo- and hypertension, arterial stiffening, heart disease, arrhythmia and so forth.

Bur et al (2000) studied the accuracy of one commercial (Hewlet Packard) oscillometric blood pressure monitor. They tested three different cuff widths on their subjects, who were critically ill patients, using intra-arterial (radial artery) pressure as a reference. The authors concluded that the device significantly underestimated arterial blood pressure and produced a
high number of measurements out of the clinically acceptable range. Therefore, this type of oscillometric blood pressure measurement fails to achieve an adequate accuracy level with critically ill patients. [22]

Reeben and Epler (1973 and 1983) were the first to suggest using a modified oscillometric measurement system, where cuff pressure changes beat-to-beat on the basis of the actual mean pressure in the artery [23]. In 1996, Jagomägi et al presented a continuous differential oscillometric blood pressure method, which was capable of continuously measuring mean arterial pressure [24]. The device (UT9201) used two servo-controlled cuffs around two adjacent fingers, one for mean pressure and another for slowly changing the pressure between a value higher and lower than the mean. Jagomägi’s measurements were carried out on young volunteers using the volume clamp method (Finapres) as a reference. In many of the measurements, variations in mean arterial pressure were very similar. The achieved accuracy for mean blood pressure was (-1.1 ± 5.5) mmHg at rest, (0.5 ± 6.9) mmHg in head-up tilt and (-3.6 ± 7.7) mmHg during deep breathing. Raamat et al (1999) studied maximum oscillation criteria and found that, with contracted finger arteries, the method gives overestimated values of up to 19 mmHg [25]. With relaxed arteries, however, the error is less evident. In their next papers (2000), Raamat et al tested accuracy during local hand heating and found no statistical error in the absence of vasoconstriction (0.3 ± 0.3 mmHg) [26]. With peripheral vasoconstriction, on the other hand, there was a statistically significant difference (6.7 ± 2.0) mmHg). Raamat et al (2001) also studied whether accuracy is affected by cooling the arm locally; their results were statistically significant ((-1.5 ± 1.1) mmHg before cooling and (8.8 ± 6.3) mmHg after cooling) [27]. Jagomägi (2001) recommended that intensive vasoconstriction, which can be determined by a laser-Doppler skin flow meter, should be avoided during measurements[28].

Volume clamp, or vascular unloading, method

Penaz invented the continuous noninvasive finger blood pressure measuring method based on a transparent finger cuff, photoelectric plethysmogram, i.e. transmitter and detector, and pressure controller unit in 1973. Molhoek et al (1984) then tested it with 21 hypertensive patients using intra-arterial pressure as a reference [29]. They obtained an underestimation of 6 mmHg for both systolic and diastolic blood pressure, making the method attractive for further analysis. Gravenstein et al (1985) studied tissue hypoxia in order to assess its impairing effect on blood circulation in the finger tip on the distal side of the finger cuff [30]. They observed an oxygen desaturation from 97% to 93.7%, recorded after applying cuff pressure for one minute. Boehmer (1987) from Ohmeda Monitoring Systems introduced a prototype device, later known
as Finapres™, based on the vascular unloading method, in which the cuff on the finger is pressurized to the same pressure as the artery [31]. Cuff pressure was controlled by electronic, automatic adjustments so that it equalled the intra-arterial pressure at all times and pulsed with arterial pressure. These measurements consisted of photoplethysmographic measurements of finger volume; a light source and a detector located inside a transparent cuff measured blood volume alterations, i.e., blood flow. Cuff pressure was controlled by servo technique using a microprocessor to keep the blood flow stable; when the pressure inside the artery increased, external cuff pressure increased by the same amount at the same time. The system automatically corrected changes induced by smooth muscle contractions or relaxations.

Factors affecting Finapres’ reliability and optimal measurement conditions were studied by Kurki et al (1987) [32]. They measured 50 men during surgical operations using four different fingers and three different types of cuff and established that, compared to the intra-arterial blood pressure, the highest accuracy was achieved with the thumb: the observed accuracies were -4.8 mmHg for systolic, 1.49 mmHg for diastolic and 0.29 mmHg for mean arterial pressure. The correlation coefficients were 0.945, 0.884 and 0.949, respectively, indicating high accuracy. Epstein et al (1989), however, studied obstetric patients during spinal anaesthesia and compared Finapres’ accuracy with that of other automated oscillometric methods [33]. Differing greatly from previous studies, their results indicated an accuracy of (6.6 ± 12.5) mmHg for systolic, (3.3 ± 10.4) mmHg for mean and (7.2 ± 9.8) mmHg for diastolic pressure. Later Kugler et al (1997) ran a series of tests with a sphygmomanometer and concluded that there were systematic differences in the results of the different methods [34]. It must be pointed out, however, that their reference method was noninvasive and thus less accurate than the invasive method used by Kurki et al. Shortly after, Kurki et al (1989) published a study on open-heart surgery, which demonstrated that Finapres worked well with 13 patients out of 15, before they had a cardiopulmonary bypass operation and in 10 out of 15 during the operation [35]. They reported average accuracies of (8.3 ± 10.2) mmHg and (6.6 ± 8.7) mmHg with the correlation coefficients of 0.814 and 0.902, respectively. Since then, a number of papers have explored the accuracy of Finapres. Bos et al (1996) and Wesseling (1996) have also defined a specific correction method for Finapres and Portapres (Shmidt 1991), in which the arterial pressure waveform recorded on a finger is filtered and bias-corrected to closely correspond to intra-brachial pressure. To conclude this discussion, it must be pointed out that, as a result of extensive combined studies conducted by Silke & McAuley (1998) and Imholz et al (1998, the accuracy of Finapres is now broadly acknowledged. More than 150 papers have been published on blood pressure measurements using Finapres, some of which are tabulated in Table 3.
Table 3. Comparison of the accuracy (mean values ± standard deviations) of the Finapres (also Portapres and Finometer) and intra-arterial pressure (IAP). [36]

<table>
<thead>
<tr>
<th>Year</th>
<th>Authors</th>
<th>Reference method</th>
<th>Num.</th>
<th>Comments</th>
<th>Systolic accuracy (mmHg)</th>
<th>Mean accuracy (mmHg)</th>
<th>Diastolic accuracy (mmHg)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1991</td>
<td>Schiller Z. &amp; Pash T.</td>
<td>IAP</td>
<td>I: 15</td>
<td>Two groups, satisfactory reliable</td>
<td>I: -1.6 ±16.6, II: -2.7 ±13.1</td>
<td>I: 0±2.6 II: -2.6 ±10.9</td>
<td>I: 4.6 ±11.5 II: 4.7±9.8</td>
</tr>
<tr>
<td>1991</td>
<td>Epstein R. H. et al</td>
<td>IAP</td>
<td>Not reliable during anaesthesia</td>
<td>12</td>
<td>Hypertensive and vascular disease</td>
<td>5.8±11.9</td>
<td>7.7 ±10.0</td>
</tr>
<tr>
<td>1991</td>
<td>Bos W. J.</td>
<td>IAP</td>
<td>15</td>
<td>Elderly at rest</td>
<td>-15.7±18.8</td>
<td>-20.1±15.7</td>
<td>-13.5±15.7</td>
</tr>
<tr>
<td>1992</td>
<td>Novak V. et al</td>
<td>IAP</td>
<td>20</td>
<td>One hour recording</td>
<td>0.84 ±13.3</td>
<td>6.67 ±5.23</td>
<td>-13.4</td>
</tr>
<tr>
<td>1994</td>
<td>Rongen G. A. et al</td>
<td>IAP</td>
<td>15</td>
<td>Operated pediatrics</td>
<td>-16.8±2.6</td>
<td>-10.8±1.5</td>
<td>-17.5±1.6</td>
</tr>
<tr>
<td>1995</td>
<td>Bos W. J. et al</td>
<td>IAP</td>
<td>28</td>
<td>Healthy elderly and vascular diseased patients</td>
<td>-3.2±16.7 corr.:</td>
<td>-3.7±7.0</td>
<td>-0.7±4.6</td>
</tr>
<tr>
<td>1996</td>
<td>Silke B. &amp; McAuley D.</td>
<td>IAP</td>
<td>449</td>
<td>Combined databases</td>
<td>-0.8±11.9</td>
<td>-1.6±7.6</td>
<td>-1.6±8.3</td>
</tr>
<tr>
<td>1998</td>
<td>Imholz B. P. M. et al</td>
<td>IAP</td>
<td>1031</td>
<td>Combined databases</td>
<td>2.2±12.4</td>
<td>2.1±8.6</td>
<td>0.3±7.9</td>
</tr>
<tr>
<td>2002</td>
<td>Westehof B. E.</td>
<td>IAP</td>
<td>14</td>
<td>Hypertensive, 24 h, Portapres, correction</td>
<td>1±10 corr:</td>
<td>-10±8</td>
<td>-8±7</td>
</tr>
<tr>
<td>2003</td>
<td>Guelen I. et al</td>
<td>IAP</td>
<td>37</td>
<td>Age 41-83, Finometer, 1st corr.</td>
<td>-9.7±13.0 corr:</td>
<td>-1.1±10.7 corr:</td>
<td>-1.5±6.6</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td></td>
<td>2nd corr.</td>
<td>3.1±7.6</td>
<td>2.7±4.7</td>
<td>4.0±5.6</td>
</tr>
</tbody>
</table>

Finapres, Portapres and Finometer are widely used in different types of physiological measurements. For example, in 60 degree head-up tilt tests (Friedman 1990), the obtained blood pressure variation correlation coefficient was 0.98 with intra-arterial blood pressure. [37]

Finapres devices are no longer available, since they have been replaced by another device, known as Finometer, a trademark of FMS, Finapres Medical Systems BV, Arnhem, Netherlands. The company also produces PortapresTM blood pressure monitors, which measure two fingers alternately, enabling one finger to normalize while the other is being measured. Langewouters et al (1998) encourage physicians to use Finapres or Portapres instead of invasive methods [38].
**Tonometric method**

First presented by Pressman & Newgard in 1963, the arterial tonometer is a pressure measurement method that can noninvasively and continuously record pressure alterations in a superficial artery with sufficient bony support, such as the radial artery. It uses a miniature transducer (e.g., by Millar Instruments) or a rigid sensor array (e.g., by Colin Electronics) or a flexible diaphragm (Drzewiecki et al 1983), which is attached on the skin above the pulsating artery [39]. Skin and tissue located between the sensor and the array transfer pressure pulsations between them. When the pulsations reach their strongest level, the sensor is regarded as being correctly positioned. This can be facilitated by using a sensor array and selecting sensor elements with the strongest amplitude. This method requires that the sensors are closely alike in terms of sensitivity. Next, the sensor or sensor array is pushed towards the vessel using, for example, air pressure. The vessel flattens when the pressure and, consequently, the force against the artery wall increases. Arterial pressure in the top of the flattened artery’s center equals the supporting pressure, allowing the recording of an accurate blood pressure profile. If the pressure increases too much, the artery will occlude totally and the measurement will be erroneous. Thus, the hold-down pressure must be continuously controlled using stress distribution information. Furthermore, the size of the sensor or sensor element must be small relative to the artery and the sensor material must be rigid, which has led to the use of piezoelectric or piezoresistive (made of silicon) materials. Moreover, sensor arrays enable the use of motion artifact cancellation methods to improve the signal-to-noise ratio. The measurement method and the piezoresistive silicon sensor array were presented by Weaver et al in 1978 and by Terry et al in 1990 [40, 41]. Developed in the USA in co-operation with SRI International & IC Sensors, the sensor has been marketed by Colin Electronics, Co., Ltd., Japan (Ohmori 1992) since 1992. Zorn et al (1997) validated the tonometer monitor, known as Colin Pilot 9200, with 20 adult and paediatric patients using intra-arterial blood pressure as a reference [42]. They achieved an accuracy of $(2.24 \pm 8.7) \text{ mmHg}$ and $(0.26 \pm 8.88) \text{ mmHg}$, which slightly exceeded the allowable standard deviation of error. Since then, the Colin tonometer has been used to estimate the waveform of central aortic blood pressure (Hori et al 1997, Chen et al 1997 [43], Fetics 1999 [44]) using mathematical transformation, i.e., a transfer function between a noninvasive tonometer and an invasive catheter measurement site. An autoregressive exogenous model of the Colin tonometer signal proved accurate in studies by Hori (approximately a 3% estimation error for systolic, mean and diastolic pressure) and Fetics ($(0.4 \pm 2.9) \text{ mmHg}$). Chen studied a generalized transfer function model and obtained an almost identical reading ($(0.2 \pm 3.8) \text{ mmHg}$). Nowadays, the Colin Pilot is represented by DRE Medical, Inc., USA.
Another type of tonometer is based on volume changes in a fluid-filled, flexible diaphragm chamber (Drzewiecki 1995 [45]). In this setup, the sensor base is made of plexiglass. A rounded rectangle-shaped channel is machined into it, and the sensor array is sealed with a thin polyurethane sheet, functioning as a flexible diaphragm. Next, the channel is filled with saline. At each end of the channel, there are two stainless steel electrodes used to inject a current along the channel length. Near the centre of the channel, four measuring electrodes are placed at equal spacing. Each pair of electrodes defines a volume compartment and the voltage across each pair is calibrated in terms of volume using impedance plethysmography. External to the tonometer, a saline-filled catheter is used to connect the channel to an electro-mechanical volume pump. A Velcro strap can be used to fix the tonometer in place. When the tonometer is attached on the skin over a superficial artery, such as the radial artery, blood pressure pulsations in it produce volume shifts inside the volume sensor resulting in an impedance change. The pressure inside the tonometer can be varied so that the strongest signal can be measured. This pressure corresponds to mean arterial pressure, MAP.

**Pressure pulse transit-time method**

Galen (131-200 AD) was the first in Europe to note the relationship between the heart and the arteries, and he also pointed out that the arteries contain blood, not air. He ranks as the foremost sphygmologist of Antiquity, not merely because he wrote a number of books on the subject, but because his teaching on the pulse dominated clinical practice for about 16 centuries. The first attempt to represent the pulse graphically, in a form comparable with sphygmographic tracing, was made in 1540 by Struthius, who studied pulse waves by placing a leaf on the radial artery and watching its vibrations. It was only in 1553 that Miguel Serveto described pulmonary circulation, and in 1628 that William Harvey provided the first description of blood circulation. Harvey’s work established the pulse wave as a manifestation of cardiac ejection. More than 100 years later, in 1731, the Reverend Stephen Halens recorded the first invasive arterial pressure measurement. Etienne Marey, in turn, was the first to accurately record arterial pulses in humans using a sphygmograph. Later, several sphygmographic devices were devised. [46]

It was recognized quite early that the elasticity of an artery is related to the velocity of the volume pulses propagating through it. Moens (1878) and Korteweg (1878) derived a mathematical expression for the velocity of the pulse front traveling along an artery as a function of such factors as the elasticity coefficient, the thickness of the arterial wall and the end-diastolic diameter of the vessel lumen. This is shown in the formula below, where $PWV$ is pulse wave velocity; $t$ the thickness of the vessel wall, $d$ the diameter of the vessel, $\rho$ the density of blood and $E$ stands for Young’s modulus describing the elasticity of the arterial wall. [46]
Pulse transmission velocity was used as a clinical index of arterial elasticity by Bramwell and Hill (1922), and later by a large number of authors using different methods. In 1981, Geddes showed that diastolic blood pressure can be monitored continuously using extravascular pulse pickups and time-domain techniques to determine the pulse-wave velocity.

Being dependent on pressure, Young’s modulus \( E \) is not a constant. The relationship is of the form \( E = E_0 e^{\alpha P} \), where \( E_0 \) is the zero pressure modulus, \( \alpha \) is a constant that depends on the vessel, and \( e \) is 2.71828 (Geddes 1991, reading 100-103). Thus, Formula (1) can be expressed as

\[
PWV = \sqrt{\frac{tE}{\rho d}}.
\]

Geddes et al proposed in 1981 a method where blood pressure pulses will be measured on the same artery in different locations along the vessel. In this method, the pulses become detectable, when the cuff pressure applied on the upper arm falls below systolic blood pressure. Moreover, as the pressure decreases, the pulse delay shortens until it stabilizes. This point can be used to determine diastolic blood pressure. Geddes used this method to measure anaesthetized dogs. [47]

In his paper of 1988, Okada showed that PWV correlates well with age, gender, systolic and diastolic blood pressure and serum phospholipids [48]. He used two sets of photoplethysmographs to record pulses in the fingertip and toe tip. Later in 1996, Franchi et al applied the ECG, a direct blood pressure measurement and photoplethysmography [49], PPG, to the ear lobe. Using the first and second derivative of the photoplethysmographic signal to determine pulse onset, they found a correlation between delays from the R-wave to the aortic pulse and from the aortic pulse to the ear lobe pulse.

Since then, pulse wave velocity measurements have attracted a great deal of attention. During the 1990s and 2000s, it has become a broadly approved method of assessing arterial stiffness, which has proven to be a good marker for predicting premature death. In 1999, Dr. Roland Asmar wrote a book which summarizes the principles of the method, its clinical applications and factors influencing pulse wave velocity. In principle, pulse waves can be measured noninvasively on the skin over the artery between two different points in the arterial tree. As seen in the formula above, the elastic modulus is dependent on pressure, allowing it to be determined on the condition that we can measure pulse wave velocity. The modulus alternates in different positions in the arterial tree depending on the vascular tonus, i.e.
vasoconstriction and vasodilation, which makes it difficult to measure PWV accurately. In his book, Asmar also mentioned that pulse wave velocity depends on age, gender, genetic factors, salt intake, heart rate and blood pressure.

Photoplethysmography for the pulse transit-time method

Since the discovery of the Penaz -Volume clamp method and experience gained of the Finapres blood pressure measurement method, optical photoplethysmographic, or PPG, finger pulse detection has been the principal pulse detection method for many researchers. In 2000, Chen et al proposed a method based on measuring the beat-to-beat time interval between the QRS apex in the ECG signal and the onset of photoplethysmogram in an oximeter sensor placed on a fingertip [50]. The method was tested with 20 cardiac patients during a surgical operation using an invasive method as a reference. PTT values were calculated and divided into two bands, high (HF, 0.005-0.04 Hz) and low (LF, DC-0.005 Hz), by digital filtering. Before that, the time scale was reorganized such that one blood pressure pulse corresponded to one PTT value. The method used invasive blood pressure recordings for intermitted calibration (systolic DC-level). The estimated values for systolic blood pressure were calculated by summing HF- and LF-values, and the obtained accuracy for systolic blood pressure was 

Two research groups, one led by Prof. Asada at the Massachusetts Institute of Technology, USA, and the other by Prof. Zhang in Hong Kong, have explored the topic since the turn of the millennium. Their main results on blood pressure measurements will be presented in the next paragraphs.

Starting with Asada’s research group; Yang et al and Rhee et al introduced the ring sensor in 1998. It comprised a photoplethysmography and a pulse oximetry monitor in a ring, which could be attached around a patient’s finger for continuous measurements 24 hours a day. In addition, the device contained a microcontroller (PIC16C11) required to control the pulsing of the LEDs (NIR, 940 nm and red, 660 nm), data acquisition, signal conditioning and filtering. It also had a MEMS (micro electro machined system) accelerometer for artifact rejection and an RF transmitter (OOK, on-off-keying) and a button-type lithium battery.

In the next generation of devices (Rhee et al 2001 and Asada et al 2003), this single ring sensor arrangement was replaced by a double, isolating, ring system, where the inertial mass of the battery and printed circuit board could not influence the inner ring, where the LEDs and the photo diode were. This system was artifact-resistant and insensitive for ambient lighting,
enabling its application to moving, continuous 24-hour heart rate monitoring for almost a month using a button cell battery. In addition, with a discrepancy of only 1.23 pulses/min in terms of rms values, it provided good accuracy.[51, 52]

In 2005, Gibbs & Asada proposed an artifact recovery method, based on a MEMS acceleration sensor. The method subtracts the signal produced by the MEMS sensor from the corrupted PPG sensor signal with a certain delay, with the result that the output exhibits a low distortion level. Although this correction method can be used either in an additive or logarithmic model, the latter proved better, and it can also be enhanced to include the Laguerre expansion technique (Wood & Asada 2005).

Shaltis and Asada et al in 2005 designed a new version of the photoplethysmography sensor for measuring five vital signs of a patient, including temperature monitoring. This new device consists of a LED array and a photo diode array to locate digital arteries. Also, noisy PPG signals can be recovered by a 3-dimensional MEMS accelerometer and active noise cancellation. All data is transmitted by an IEEE 802.15.4 compliant transceiver. They also considered the possibility of using PPG signals to determine blood pressure, but found no stationary relationship between PPG and arterial blood pressure over time scales greater than 20 minutes.

Chan et al, members of Zhang’s group, presented in 2001 a method for noninvasive and cuffless blood pressure measurements for telemedicine purposes. In this method, ECG was used as a timing base and start signal, and PPG from a finger as a stop signal. The device included a WAP-based Internet connection. After 20 trial measurements, they achieved an accuracy of (7.5 ± 8.8) mmHg for systolic, (6.1 ± 5.6) mmHg for mean and (4.1 ± 5.6) mmHg for diastolic blood pressure using an unspecified commercial cuff measurement as a reference.

Sometime later, in 2003, Zheng & Zhang published a paper describing measurements with three different types of PPG ring: convex, flat and concave. A LED and a detector were fixed on the palm-side of the ring, while the other side housed two ECG electrodes, one on the inside surface and the other on the dorsal side. A convex-type sensor, in which the LED and the detector are 1 cm apart on the palm side, proved best in compensating for motion artifacts. Lee & Zhang also proposed mathematical methods to compensate for motion artifacts, referred to as stationary wavelet transform and wavelet transform modulus maxima, achieving a remarkable improvement. Teng & Zhang used this PPG ring for blood pressure measurements on 15 young, healthy subjects. To provide reference measurements, they used an oscillometric BP-8800 device manufactured by Colin, Ltd. Measurements, taken at three different stages: rest, step-climbing exercise and recovery from the exercise, indicated an accuracy of (0.2 ± 7.3) mmHg for systolic and (0.0 ± 4.4) mmHg for diastolic blood pressure.
A year later, Hung et al reported a wireless measurement concept using Bluetooth for telecommunication from a PPG ring to a mobile phone, PC or PDA device. The achieved accuracy was \((1.8 \pm 7.6)\) mmHg for systolic and \((0.5 \pm 5.3)\) mmHg for diastolic blood pressure.

Poon & Zhang extended the measurements to include 85 subjects aged 57 (average) \(\pm 29\) (min, max) years, of whom 36 were males and 39 hypertensives. These measurements, conducted over an average period of 6.4 weeks, resulted in about one thousand pairs of systolic and diastolic blood pressure estimations being made. Reference results were provided by the average of results measured by a nurse using the auscultatory method and results obtained using two clinically approved automated blood pressure meters (BP-8800 Colin, Ltd and HEM-907, Omron, Ltd). Poon & Zhang reached an accuracy of \((0.6 \pm 9.8)\) mmHg for systolic and \((0.9 \pm 5.6)\) mmHg for diastolic blood pressure. [53]

**Electronic palpation method**

Pressure array transducers based on EMFi-material have been produced, tested and used to measure the heartbeat rate of immobile and moving persons [54]. In addition, they have been utilized in non-invasive blood pressure measurements on healthy volunteers and cardiac patients. In these measurements, a cuff was applied to the upper arm and pulsations were sensed on the distal side of the wrist. Essential findings have already been published in several conference proceedings, journals (Sorvoja et al 2003-2006, [55-58]) and in Doctoral Thesis [36].

In blood pressure measurements of this kind, based on the electronic palpation of the amplitude and the transit time of pressure pulse waves, pulsations are detected by a wrist array transducer during the inflation or deflation of the cuff. In the deflation mode, cuff pressure is rapidly inflated over the presumed systolic pressure level and then slowly deflated under the diastolic level. The onset of pulsations corresponds to systolic blood pressure. Diastolic blood pressure, on the other hand, can be determined by the point where the delay in the pressure pulse sensed by the transducer array stops to diminish. Though the same method can be used during cuff inflation or deflation, the latter mode is preferable.

The measurement device used in first measurement set consisted of a standard 13 cm cuff, a wristwatch-type four-channel pressure transducer array, an invasive catheter (Ohmeda), an amplifier, a filtering unit and an automatic pressure-controlling unit. A lap-top computer equipped with a data acquisition board by National Instruments (NI Daq 700) was used to measure the outputs of the amplifiers. Signals were sampled at the 100 Hz frequency. Based on electro-thermo-mechanical film, the transducer array was specifically designed for detecting pulsations in the radial artery. The amplifier unit was used to amplify and band-pass filter cuff
pressure signals (1-15 Hz, Butterworth filter response, fourth order high pass, second order low pass) and signals produced by the sensor.

The next measurements setup was based on signals produced by the DATEX™ patient monitoring system and another, self-made, blood pressure measurement device. In addition to the device described above, this one also comprised an MSP430™-based microcontroller unit for determining blood pressure. This system is illustrated in Fig. 1, with the connection cable to the patient monitoring system missing.

Fig. 1. Second measurement system. From left to right: self-controlled pressure pump unit, amplifier and band-pass filter unit, MSP430-microcontroller unit, cuff for adults with an arm circumference of 24-33 cm (Speidel & Keller), aneroid pressure metering device with a pump for pressure calibration and lap-top computer equipped with the National Instrument DAQ card 700. In the lower right hand corner is the four-element wrist sensor used to detect pressure pulses in the radial artery.

Using 64 kBytes, the measurement and blood pressure determination algorithm was coded in the microcontroller manufacturer’s assembly language (~1900 rows text). The compiled program used 512 bytes of RAM and 3970 bytes of ROM. Cuff pressure was inductively transmitted to the microcontroller unit by transmitter-receiver pairs designed by Polar Electro Ltd. Pressure information was based on time-interval measurements, with one pulse a second corresponding to zero pressure and two pulses a second to a pressure of 256 mmHg. Also the sensor’s signal band was slightly different from that of the first measurement system: four-element transducer signals were amplified and filtered using a first-order band-
pass filter (1.7 Hz-11 Hz) in a full custom integrated circuit. A connection from the DATEX\textsuperscript{TM} device provided an ECG signal and each patient’s radial and pulmonary intra-arterial blood pressure. An automatic pressure-controlling unit, connected to the laptop computer, started cuff pressure inflation and sent all pressure data wirelessly to the microprocessor unit. Designed to assess the functionality of the whole system, the microprocessor unit contained the same blood pressure detection algorithm as the laptop computer. The data acquisition rate of the recording system was 100 samples/second.

**Study subjects**

The first set of measurements was carried out at the Oulu University Hospital. Most of the study subjects, 16 in all, were healthy volunteers. In addition, seven elderly patients were measured. It soon became apparent that the blood pressure signals of the two groups differed in shape and amplitude. At that time, the blood pressure determination algorithm focused on detecting changes in pulse amplitude, not in the recorded delays near the diastolic pressure level. The results of measurements made on healthy persons, and their characteristics are shown in Table 4 (average values and standard deviations). An invasive catheter (Ohmeda) was used as a reference blood pressure method. These measurements were made twice, in supine and sitting position, during inflating and deflating cuff pressure with a one-minute follow-up recording period between them. It can be seen that both the men and the women were normal weight, and that their blood pressures were in the normal or high normal range. The width of the cuff was 13 cm.
Table 4. Study subjects (healthy volunteers). The values are average values ± standard deviations or range [57].

<table>
<thead>
<tr>
<th>Item</th>
<th>Men</th>
<th>Women</th>
</tr>
</thead>
<tbody>
<tr>
<td>n</td>
<td>12</td>
<td>4</td>
</tr>
<tr>
<td>Age, years</td>
<td>26.2 (20–33)</td>
<td>30.8 (27–34)</td>
</tr>
<tr>
<td>Height, cm</td>
<td>180.8 ± 5.6</td>
<td>170.0 ± 2.4</td>
</tr>
<tr>
<td>Weight, kg</td>
<td>79.4 ± 10.1</td>
<td>63.5 ± 6.0</td>
</tr>
<tr>
<td>BMI, kg/m²</td>
<td>24.2 ± 2.4</td>
<td>22.0 ± 1.9</td>
</tr>
<tr>
<td>Arm circumference, cm</td>
<td>31.6 ± 2.6</td>
<td>28.3 ± 2.4</td>
</tr>
<tr>
<td>Blood pressure measurements, n</td>
<td>98</td>
<td>23</td>
</tr>
<tr>
<td>IA systolic BP, mmHg</td>
<td>129.1 ± 11.2</td>
<td>122.3 ± 10.1</td>
</tr>
<tr>
<td>IA mean BP, mmHg</td>
<td>87.3 ± 6.5</td>
<td>83.5 ± 5.7</td>
</tr>
<tr>
<td>IA diastolic BP, mmHg</td>
<td>67.4 ± 5.0</td>
<td>63.7 ± 5.3</td>
</tr>
<tr>
<td>Heart rate, bpm</td>
<td>66.2 ± 11.3</td>
<td>69.8 ± 8.1</td>
</tr>
</tbody>
</table>

The next set of measurements was conducted on elderly cardiac patients at the Oulu University Hospital. Also in these measurements, reference readings were provided by the invasive method with the signal obtained in analog form directly from the patient monitor’s output.

In the end, three to five measurements on 51 patients were saved on the computer for further analysis. Unfortunately, a number of measurements had to be rejected due to inadequate recording (missing arm circumference) or insufficient cuff width. As the cuff width in all measurements was 13 cm, the study subjects’ arm circumference had to be within 24-33 cm to ensure adequate accuracy (Geddes 1991, manufacturer’s claim). Table 5, tabulating the characteristics of these eligible patients, shows that their arm circumference was appropriate and that the invasive blood pressure values were within the optimal normal range.

Table 5. Test subjects (cardiac patients). To provide a single value for each patient, the intra-arterial (IA) blood pressure (BP) values given are average values of the measured IA diastolic, mean and systolic BP values (average values ± standard deviations or range). [57]

<table>
<thead>
<tr>
<th>Item</th>
<th>Men</th>
<th>Women</th>
</tr>
</thead>
<tbody>
<tr>
<td>Number</td>
<td>14</td>
<td>6</td>
</tr>
<tr>
<td>Age, years</td>
<td>67.8 (59–75)</td>
<td>65.2 (49–75)</td>
</tr>
<tr>
<td>Height, cm</td>
<td>168.0 ± 8.1</td>
<td>163.0 ± 6.6</td>
</tr>
<tr>
<td>Weight, kg</td>
<td>71.4 ± 11.8</td>
<td>74.1 ± 10.0</td>
</tr>
<tr>
<td>Body mass index, kg/m²</td>
<td>25.4 ± 4.4</td>
<td>28.0 ± 4.6</td>
</tr>
<tr>
<td>Arm circumference, cm</td>
<td>32.3 ± 1.0</td>
<td>32.7 ± 0.5</td>
</tr>
<tr>
<td>Bypass operation</td>
<td>13</td>
<td>4</td>
</tr>
<tr>
<td>Valve operation</td>
<td>3</td>
<td>2</td>
</tr>
<tr>
<td>Blood pressure measurements, n</td>
<td>40</td>
<td>13</td>
</tr>
<tr>
<td>IA systolic BP, mmHg</td>
<td>114.8 ± 21.3</td>
<td>118.3 ± 19.9</td>
</tr>
<tr>
<td>IA mean BP, mmHg</td>
<td>77.8 ± 9.2</td>
<td>79.6 ± 14.8</td>
</tr>
<tr>
<td>IA diastolic BP, mmHg</td>
<td>59.6 ± 6.1</td>
<td>59.6 ± 14.5</td>
</tr>
<tr>
<td>Heart rate, pulses/min</td>
<td>87.2 ± 13.5</td>
<td>81.5 ± 5.4</td>
</tr>
</tbody>
</table>

Blood pressure determination in the electronic palpation method
Systolic blood pressure can be easily determined at the point where the last (or first with deflating cuff pressure mode) radial artery pulse appears in the wrist transducer. We used an array type of transducer to facilitate the placement of the wrist transducer, as the array element with the strongest pulsation is easy to pick out.

Surprisingly, the palpated signal does not begin to decrease after the point at which cuff pressure exceeds diastolic blood pressure. In fact, the signal amplitude slightly increases, which is why the point where the amplitude signal begins to decrease cannot be used to determine diastolic blood pressure. Consequently, diastolic pressure is determined at the point where the pulse delay begins to increase, and systolic blood pressure at the point where the last pulse is detected. The cuff pressure at these time points corresponds to diastolic and systolic pressure.

Diastolic blood pressure determination is based on pulse transit time change. When pressure in the cuff exceeds the diastolic pressure level, pulses on the distal side of the cuff will be delayed. To accurately measure the point where this delay begins requires an accurate timing base. The best timing reference is afforded by the radial artery of the other hand, and, if appropriate data processing is performed, the time difference can be easily determined.

Systolic pressure was determined visually from the last palpated pulse, while diastolic blood pressure readings were obtained by the time delay change method described above. Reference values were provided by the average IA systolic and diastolic pressure. The corresponding accuracies are presented in Table 6. For the healthy group, the measured accuracies are inadequate for fulfilling the AAMI standard. On the other hand, the deflating mode yielded better results for diastolic pressure and the inflating mode for systolic pressure. For the cardiac patient group, the mean error was (-0.6 ± 5.1) mmHg for diastolic and (0.7 ± 6.3) mmHg for systolic pressure, which are considerably better than the accuracies obtained with the healthy group and fulfill AAMI standard. This can be explained by the pulse rising time: the larger the rise time, the larger the pulse delay change, which allows a more accurate determination of diastolic blood pressure.
Table 6. Accuracy (average values ± standard deviations) of the diastolic and systolic blood pressure measurements in the healthy and the cardiac patient group using the EP method. The reference value is the average value of IA diastolic, mean and systolic blood pressure. [57]

<table>
<thead>
<tr>
<th>Mode</th>
<th>Mean errors</th>
<th>Diastolic (mmHg)</th>
<th>Systolic (mmHg)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Healthy</td>
<td>Inflating</td>
<td>Supine</td>
<td>−5.5 ± 5.2</td>
</tr>
<tr>
<td></td>
<td></td>
<td>Sitting</td>
<td>−11.2 ± 6.1</td>
</tr>
<tr>
<td></td>
<td>Inflating</td>
<td>Supine</td>
<td>2.0 ± 6.6</td>
</tr>
<tr>
<td></td>
<td>Deflating</td>
<td>Supine</td>
<td>0.1 ± 7.3</td>
</tr>
<tr>
<td></td>
<td></td>
<td>Sitting</td>
<td>−0.6 ± 5.1</td>
</tr>
</tbody>
</table>

EMFi material proved applicable to detecting blood pressure pulsations on the radial artery and can also be used for pulse shape and velocity measurements. EMFi sensor material needs an operational amplifier operating as a transducer for detection. Power consumption is not a problem, however, since it is possible to use EMFi material in conjunction with an ultra low power sensor array, such as the TLC1079 by Texas Instruments, which typically consumes only 10 µW per channel.

By fulfilling the AAMI standard, the electronic palpation method proved accurate with the elderly cardiac patient group. With the healthy volunteers, however, the method only passed the requirements when it came to deflating cuff pressure and determination of diastolic pressure. The difference of the diastolic errors can be easily explained: the cardiac patient group exhibited a slower pressure pulse slope and therefore a larger delay time change during the measurement, yielding a clear indication of the changing point.

5. CONCLUSION

This study first described noninvasive blood pressure measurements methods in detail, starting from the historical background and extending to modern methods, focusing on accuracies achieved.

A new measurement method, named electronic palpation method (EP), and a measurement system was developed and used at the Oulu University Hospital for two sets of measurements. In the second set, the measurement system consisted of a microcontroller-based test device for determining blood pressure values and a data recording measurement system for further analysis. The test device worked as intended.

The determination of diastolic blood pressure was described, and the available methods’ accuracy was studied. Using intra-arterial blood pressure as a reference, this study investigated measuring method based on pulse transit time. The test subjects comprised a group of healthy
volunteers and a group of cardiac patients, who had undergone either a bypass or a valve operation, or both.

In the healthy group, the EP method was found to provide a slightly better degree of accuracy with deflating cuff pressure. With the patient group, a mean error was about zero and a standard deviation 5-6 mmHg. The results indicate that the EP method offers a feasible alternative to the other methods, particularly because it also produces information on the delay and waveform of pressure pulses.

REFERENCES


