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An algorithm to determine the ankle-brachial pressure index using the oscillometric blood pressure measurement method

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Abstract. An algorithm enabling calculation of the ankle-brachial pressure index (ABPI) was developed. ABPI relates the blood pressure in different body limbs. Using ABPI the level of vessel occlusion can be predicted and used as a simple screening method for candidates for a more thorough clinical vascular examination. To measure ABPI, the blood pressure in body limbs must be measured. In our approach, the oscillometric blood pressure measurement method was applied. It employs the usage of an inflatable cuff placed on the measured limb. The blood pressure is measured by observing the squeezed vessel pulsations caused pressure oscillations in the cuff as a function of the absolute air pressure in the cuff. First, the pressure course is preprocessed and systolic, diastolic, mean blood pressure, heart rate and channel delays are then calculated. The algorithm was developed using 59 simulated and 13 real patients-obtained pressure courses.

Key words: oscillometric blood-pressure measurement, ankle-brachial pressure index

1 INTRODUCTION

With the changing of the lifesystle, there is an increasing risk for cardiovascular diseases [1]. Therefore a serious need for a quick and effective cardiovascular disease detection and diagnostic method is present [2]. A partial blood-vessel occlusion in body's limbs is one of the most frequent vascular diseases. The direct method currently used in diagnosing the disease is the vascular ultrasound examination. Using an ultrasound examination device, a properly trained specialist can identify the location and degree of occlusion in the affected blood vessel [3]. Such examination is time consuming and complex, requiring a trained medical specialist using relatively expensive equipment, usually unavailable in a general clinic. Therefore, the examination is expensive and consequently out of reach to broad population. Our goal was to develop a screening diagnostic method, available to broad population.

1.1 Ankle-brachial pressure index computation

Our indirect diagnostic method enabling fast detection of partial blood-vessel occlusion in limbs is based on measurement of the ankle-brachial pressure index (ABPI). ABPI is the relation between blood pressures measured in different body limbs, respectively arm and leg. It can be used as an indicator of an occluded blood-vessel in a particular limb. If a blood-vessel in a limb is occluded, the blood pressure after occlusion drops [4]. The main idea of the proposed diagnostic method is to measure the concurrent blood pressure in different limbs and to compare the measured values. When the values differences are significant, the patient is a candidate for a more thorough examination.

Since it is relatively simple to measure the blood pressure in a body limb, compared to a vascular ultrasound examination, there is no need for highly trained specialist. Moreover, the blood pressure measurement requires relatively cheap equipment compared to the ultrasound examination device. The examination can be performed in a general clinic. The method can therefore be used as a screening method for identifying candidates for a thorough ultrasound examination.

2 PROBLEM DEFINITION

The main focus of our work was on development of an ABPI computation algorithm. The algorithm runs within a device which concurrently measures the blood pressure in three limbs - one hand and both feet. The measurement is performed using the oscillometric blood-pressure measurement method in which the blood pressure is measured by measuring the air pressure in a cuff, fitted to a limb. Blood-pressure oscillations in the squeezed blood-vessels produce measurable air-pressure oscillations in the cuff. The algorithm determines the level of these oscillations relative to the ciff air pressure.

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Based on this, the systolic, diastolic and mean values of the blood-pressure in the limb are estimated. Using blood-pressure values measured in each limb, APBI is calculated. Once APBI is known, the degree of blood-vessel occlusion can be estimated and the patient can then be classified as a potential candidate for a thorough examination using direct diagnostic methods.

3 Background

3.1 Blood-pressure measurement

The blood-pressure in vessels is not constant. It oscillates from its maximal - *systolic* to the minimal - *diastolic* value. It is caused by the heart muscle beating. To explain the blood-pressure course, the heart-muscle beating cycle is divided into two phases, namely systole and diastole. During the systole phase, the heart muscle contracts and forces the blood into the arteries. During the systole phase, the blood pressure reaches its highest value - the systolic blood pressure. During the diastole, the heart muscle expands and draws blood from the blood-vascular system. During this phase, just before the systolic phase begins again, the blood pressure in vessels is at its lowest value - the diastolic pressure. The blood pressure is reported in unit $[mmHg]$.

The arterial blood pressure can be measured in several ways. The most accurate method is a direct measurement requiring the pressure sensor to be connected over a tubing filled with liquid to a cannula needle placed within the arteries. The method is used in clinical practice. Since it is invasive, it is not suitable for our problem solving. Besides this invasive direct method, there are noninvasive direct blood-pressure measurement methods [6].

3.2 Current state

The most widely used methods applied in indirect measuring of the blood pressure use an air inflatable cuff squeezing the limb and vessels in which the blood pressure is to be measured.

In the auscultatory method, the cuff pressure is gradually released until it reaches the value of the highest blood pressure - the systolic pressure. At this cuff air pressure, the blood starts to squirt through previously compressed blood-vessel. Blood squirting is turbulent and can be detected as pounding - the Korotkoff sound using the stethoscope. When the air pressure in the cuff is further lowered, the Korotkoff sounds are getting louder until they start to diminish. When the cuff air pressure is lower than the diastolic pressure, the cuff does not affect the arteries any more and the Korofkoff sounds vanish [5].

A variation to this method relies on the same principle of turbulent blood-flow detection performed by the Doppler ultrasound device instead of the stethoscope.

The method using the Korotkoff sounds is not suitable for our purpose, since it requires a well trained practitioner, skilled in anatomy and able to find the compressed vessel in which the Korotkoff sounds are monitored. So, the most widely used method to measure the blood pressure is the oscillometric method [6].

3.3 Oscilometric method

The oscillometric method is an indirect blood-pressure measurement method. It measures the blood-pressure through air pressure oscillations in the cuff caused by the compressed arteries pulsations. Using this method, the cuff pressure is gradually released and measured. At a certain air pressure, when the blood pressure in vessels overcomes the air pressure in the cuff and blood starts to flow through the vessels, the air pressure in the cuff starts to oscillate. After the air pressure is further reduced, the oscillation amplitude increases to a certain maximum value and then decreases until oscillations vanish. By measuring the air oscillations amplitude the systolic, diastolic and mean blood pressure can be determined.

The oscillometric method is used in most of commercial and home usage-intended blood-pressure measurement devices [6]. While there are several general descriptions of the algorithms published [7], their details have not jet been made available.

In our reviewing the literature, we have not traced any usage of the oscillometric method in three channel blood-pressure measurement following by the ankle-brachial pressure index estimation. Similary we have not come across of an algorithm to calculate the pressure-oscillations delay enabling determination of the level of blood-vessel occlusion.

4 Materials and methods

4.1 Blood-pressure measurement

Pressure courses were measured on still laying patients on both ankles and arm simultaneously using a Mesi-developed ABPI MD device [8]. The device records the pressure course by measuring the pressure-sensor voltage using a 16 bit AD converter. The used pressure sensor SLP 33A is temperature-compensated and calibrated.

4.2 Algorithm development environment

The algorithm was developed using the Matlab version 7.9.0.. Upon the completion of the development phase, the algorithm was adapted to the blood-pressure measurement device, recoded into the C code and compiled for the ARM processor.

4.2.1 Filter design: The algorithm required two low-pass filters were designed using a Matlab filter - design tool: *filter design*.

4.2.2 Data fitting: The interpolation function enabling creation of the interpolation curve through the extremes of the cuff air pressure oscillations was designed using Matlab fitting toolbox *cftool*.

4.3 Test data acquisition

The algorithm was developed using two sets of data. The first one was obtained from a simulator simulating pressure oscillations appearing when measuring the human systolic and diastolic blood-pressure values. 59 measurements were made.

The second set was obatined from real patients. 13 patients were included in our study in which their blood pressure was measured with our device. Their blood pressure was also measured using auscalatory method by the device Ohmron M6 Comfort with the ultrasonic Doppler probe Huntleigh Dopplex MD2. Measured pressure courses were stored in files along with their reference blood-pressure measurements.

5 Algorithm

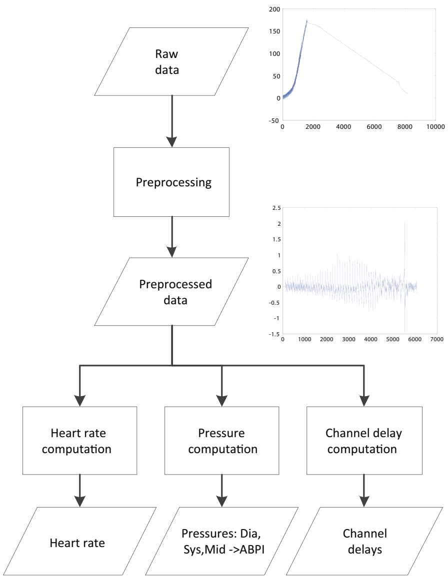

Figure 1. Algorithm overview.

The algorithm processes three channels of simultaneously measured pressure courses of three cuffs. From each pressure course it calculates the amplitude oscillation caused by vessel pulsations. The oscillations amplitude is used to determine the systolic, mean and diastolic pressure as well as the heart rate for each channel. It also calculates the pressure peaks delay between the channels.

The algorithm comprises several phases: data preprocessing, heart-rate calculation, systolic mean and diastolic pressure estimation and all three channel delays estimation. The flow of the algorithm is presented in Figure 1.

5.1 Data preprocessing

Data preprocessing provides data containing only pressure oscillations caused by compressed vessel pulsations. In this phase, digitized voltage samples form the pressure sensor are converted into actual pressure values. In the next phase, the pressure course is divided into sections representing cuff inflation and deflation. A low-pass filter is then used to eliminate the noise from the signal. The air pressure deflation course is now removed from the data in order to retain only oscillations caused by the vessel.

The algorithm starts with raw data representing the three channel pressure courses, each as a stream of voltage samples with sampling frequency $F_s = 100Hz$. Samples are quantizated using 16 bit AD converter. Figure 2 presents the unprocessed pressure data of one channel. The data is divided using a vertical line into the cuff-inflation (left) and cuff-deflation part (right). The deflation part – containing oscillations caused by the vessel pulsating is interesting

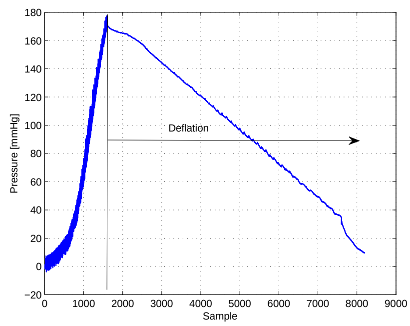

Figure 2. Unprocesed cuff-pressure course. The vertical line separates the data into the inflation and deflation part.

5.1.1 Conversion of voltage samples into actual pressure values: The relation between the cuff pressure given in units $[mmHg]$ and the voltage sampled by the AD converter from the pressure sensor is given with a set of following equations:

$$\begin{aligned} ADC_{ref} &= 2521 \\ ADC_0 &= 325 \\ n &= \frac{ADC_0 * 65536}{ADC_{ref}} \\ k &= \frac{(ADC_{200} - ADC_0) * 65536}{200 * ADC_{ref}} \\ p &= \frac{adc - n}{k} \end{aligned} \qquad (1)$$

Eqs. 1 are used to convert all the AD converter values to actual pressure values which are then processed by the algorithm.

5.1.2 Determination of inflation/deflation part: In this phase, the algorithm divides the pressure course into the inflation and the deflation part of the cuff. The pressure data course contains the whole pressure course, including pump operation part in which the cuffs are inflated to the desired pressure level. When the air pressure for each cuff is reached, pumping that cuff stops. After the required air pressure in each cuffs has been reached, air-release vales open and air pressure in the cuffs starts to decrease. The course of the deflation part is an interesting one. The exact moment of valve opening is known and synchronized for all three channels. The algorithm can therefore discard the inflation course part and keeps the deflation course part of the data.

5.1.3 Determination of the pressure course caused by cuff deflation: In this phase, the course of air pressure in the cuff caused by air deflation is determined. The course would be obtained if the cuff would be put onto an object without pulsating blood-vessels. By substracting obtained course from the pressure course, only pressure oscillations caused by the pulsating vessels are obtained. To obtain the deflation pressure course, the pressure data is filtered using a low-pass filter. We used the Hamming low-pass FIR filter of order 100 with low-pass frequency $F_c = 0,1Hz$. The frequency was chosen empirically and lies well below the expected lowest frequency of the signal representing pressure pulsations. The filter response can be observed in Figure 5.1.3.

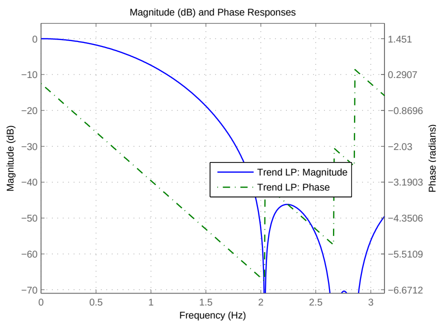

Figure 3. Low-pass Hamming FIR filter amplitude and phase response, filter order 100, $F_{cut} = 0,1Hz$. Filter is used to obtain the pressure course caused by cuff air deflation.

5.1.4 Noise removal: In this phase the high-frequency noise is filtered from the pressure signal. Noise removal made by filtering the signal using the Hamming low-pass filter of order 100 with low-pass frequency $F_c = 20Hz$. Frequency 20 Hz was chosen empirically and is well above the expected frequency of the pressure-oscillation signal. The filter response can be observed in Figure 4.

5.1.5 Elimination of the pressure course caused by the cuff air deflation: In this phase, the course caused by the cuff air deflation is eliminated leaving only pressure oscillations caused by vessel pulsations. This is achieved by subtracting the air deflation pressure course from the noise-filtered pressure course. The result of this phase can be observed in Figure 5, with only the deflation part shown. One can see, that the pressure-oscillation amplitude increases to a certain value and then decreases. The obtained signal provides the basis enabling further steps towards developing the algorithm.

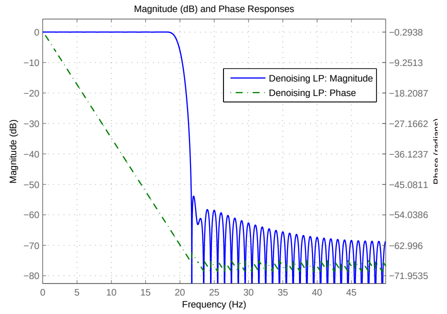

Figure 4. Low-pass Hamming FIR filter amplitude and phase response, filter order 100, $F_{cut} = 20Hz$. The filter is used to eliminate the noise from pressure signal.

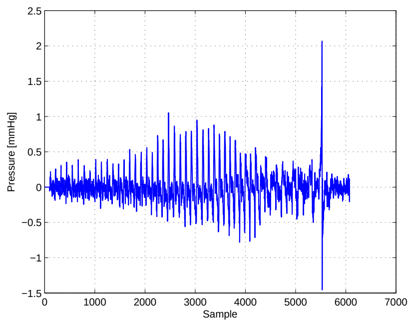

Figure 5. Cuff pressure oscillations caused by vessel pulsations. The spike at the end of the signal is an error eliminated in the next algorithm steps.

5.2 *Pressure calculation*

In this phase, the algorithm determines the absolute envelope of the pressure-oscillation signal represented in Figure 5. First, local maximums and minimums of the signal are determined enabling calculation of two interpolating functions, one for each set of extremes. The interpolating functions are then subtracted in order to obtain the function of the absolute pressure oscillation amplitude. The extreme values and functions can be observed in Figure 7. The function of the absolute pressure-oscillation amplitude of is used to determine values of the systolic, mean and diastolic blood pressures. The systolic pressure is the pressure of air in the cuff when the absolute pressure-oscillation function first exceeds certain value P_s. The diastolic pressure is the cuff air pressure when the absolute pressure-oscillation function falls below value P_d. Values P_s and P_d are determined from blood pressures measured with the reference device.

5.2.1 Local pressure-oscillations signal extreme search: To determine the envelope of the extremes they must be located in partially by the noise occluded signal [9]. Since local maximums and minimums follow each other in an alternating manner, the algorithm scans through the data alternating searching of local maximums and minimums. To declare the next local maximum, its value must differ from the previous local minimum for at least δ. The value of δ is determined empirically with regard to the analyzed data. When the algorithm finds the next extreme it records its value and index. Figure 6 shows local extreme search algorithm result.

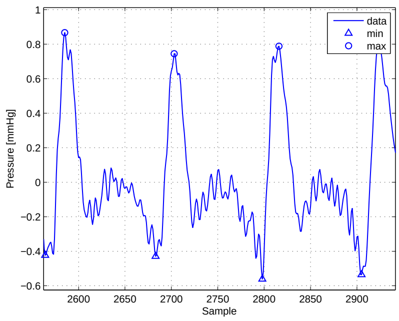

Figure 6. Local extreme search. The solid line represents the pressure course. The circles mark the found local maximums and the triangles mark the found local minimums.

By using the found local extremes interpolating function is calculated.

5.2.2 Calculation of the local pressure-oscillation extreme interpolating function: The interpolating function is calculated from local extremes using the smoothing spline interpolating function [10].

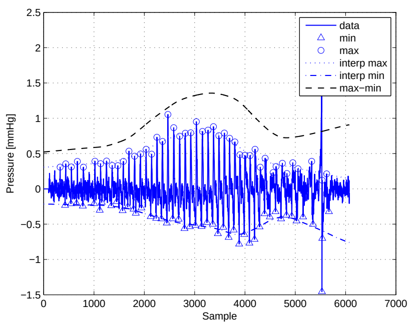

Figure 7. Pressure oscillations caused by vessel pulsating. Local extremes are found and interpolating functions are drawn through them. The difference of interpolating functions – representing absolute pressure-oscillation is also drawn.

5.3 *Calculation of ABPI*

APBI represents the ratio of pressures measured on individual limbs. It is calculated as quotient P_1/P_2.

5.4 *Heart-rate calculation*

Pressure courses caused by each heart beat are very self similar. When calculating the heart rate, the steady heart-rate during measurement is assumed .

To calculate the heart rate, the autocorrelation function of the pressure course containing the pressure oscillations is computed. When the delay of autocorrelation corresponds to exactly N x delay between the heart beats where N is an integer, the autocorrelation function peaks. The section of the autocorrelation function is shown in Figure 8. The peaks of the autocorrelation function are detected using the method described in subsection 5.2.1. The detected maximums – marked with stars can be observed in Figure 5.2.1. By using local maximums positions their mean distance t_{peak} is computed. Since sampling frequency is known ,$F_s = 100Hz$, the heart rate in beats per minute is calculated using Equation 2:

$$hr = \frac{60 * F_s}{t_{peak}} \tag{2}$$

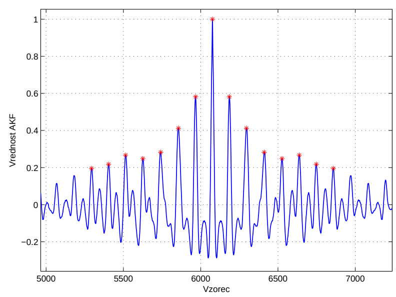

Figure 8. Section of the autocorrelation function of the pressure oscillations course measured in the cuff. The local maximums are marked with stars.

5.5 *Channel-delay calculation*

The pressure-oscillation channel delay is calculated as an additional indicator of potentially occluded vessels. Delays between channels are calculated by first calculating the correlation function between the pressure-course channels. With correlation functions computed, delays between the maximum values of the correlation functions are calculated. Figure 9 illustrates correlation functions between the channels. The delay of channels 1-3 and 2-3 is the same and that of the channels 1-2 is smaller. The sampling frequency being known, we can calculate the delay between channels d_{12}, d_{13} and d_{23} using Eqs 4:

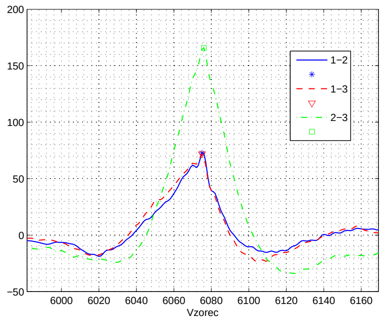

Figure 9. Pressure oscillation correlation between channels 1-2, 2-3, 1-3. Maximums on each function are marked with stars.

$$
\begin{aligned}
d_{min} &= min(idx(ch12_{max}), ..., idx(ch23_{max})) \\
d_{12} &= \frac{ch12_{max} - d_{min}}{F_s} \\
d_{13} &= \frac{ch13_{max} - d_{min}}{F_s} \\
d_{12} &= \frac{ch12_{max} - d_{min}}{F_s}
\end{aligned}
\tag{3}
$$

6 Results

By using the developed algorithm, the pressure values for the available pressure courses were calculated. The reference pressures were used to determine the P_d and P_s values. The criteria for the value determination was minimum mean-square error between the measured pressure and the calculated one. With the determined $P_d = 2.77mmHg$ and $P_s = 1.87mmHg$, the standard deviation of error for the diastolic pressure was $\sigma_d = 6.3mmHg$ and the standard deviation of error for the systolic pressure was $\sigma_s = 4.5mmHg$. ABPI values were also calculated.

7 Conclusion

The developed algorithm is ready to be used in a clinical testing on a larger number of patients with confirmed vascular diseases. ABPI values, calculated with our algorithm and known degrees of vessel occlusion will be used in building a model for estimation of the vessel occlusion. Further testing on the diseased patients will prove effectiveness of our method and help to determine which pressure should be used calculate the APBI. The testing will also reveal if the channel delay can be used in predicting the actual vessel occlusion.

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