A New Method to Determine Systolic Blood Pressure Indirectly Aided by Parallel Recording of ECG and PPG

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Abstract: Raised blood pressure severely increases the risk of lethal cardiovascular diseases. Home monitoring of blood pressure is vital in early detection and treatment of hypertonia. Accuracy of indirect blood pressure measurement methods is sensitive to many physiological factors that are difficult to measure or control. The accuracy can be improved by using further sensors. In this paper, we propose a new method for the estimation of systolic blood pressure based on cuff pressure, ECG and photoplethysmographic (PPG) signals. PPG is measured without hardware filtering keeping the DC-component and avoiding the problem of distorting the signal. The proposed method was validated by applying it to healthy senior and healthy young adults at rest and by making a measurement series containing mild physical exercise for healthy young adults. Results of the tests clearly show the supremacy of the new method to conventional oscillometric procedure.

1 INTRODUCTION

Hypertension is a major cardiovascular risk factor and the leading cause of death and disability-adjusted life-years worldwide (Carey & Whelton, 2018). Early detection and appropriate management of hypertension is based on blood pressure (BP) measurement, which is conventionally carried out by a healthcare professional in the office. However, out-of-office BP measurement has been shown to provide more reliable estimation of the average BP of an individual over time (Stergiou & Bliziotis, 2011). Home BP monitoring allows the detection of white-coat hypertension and masked hypertension and contrary to ambulatory BP monitoring; it does not require expensive instrumentation. For home BP monitoring, automated oscillometric devices are currently recommended, because they are non-invasive, easy to use and cheap devices, requiring little training compared to auscultatory devices (Stergiou et al., 2018). Despite its widespread use and many advantages, the oscillometric method has also major limitations. First of all, the classic method basically measures the mean arterial pressure (MAP), systolic blood pressure (SBP) and diastolic blood pressure (DBP) values are only calculated (Drzewiecki, Hood & Apple, 1994). The accuracy of the calculated SBP and DBP values is sensitive to several physiological factors including pulse pressure, anatomical position, elasticity and size of the measured artery and properties of the surrounding tissue (Tholl, Forstner & Anlauf, 2004). Increased arterial stiffness generally impairs the accuracy of oscillometric blood pressure measurement (van Popele et al., 2000). Since the introduction of the oscillometric method, several improved algorithms have been published including amplitude- and slope-based methods, model-based algorithms, methods based on neural networks and machine learning and the exploitation of extra sensor signals enabling the calculation of the pulse wave transit time (PWTT). Forouzanfar et al. presented a review article about the BP measurement based on oscillometric algorithms (Forouzanfar et al., 2015). Alghamdi et al. gave an overview of BP measurement methods based on machine learning and proposed a classification-based BP estimation method (Alghamdi et al., 2020). Despite the major progress in this field, oscillometric BP measurement techniques still have limitations as highlighted in (Alghamdi et al., 2020; Forouzanfar et

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al., 2015). In the present paper, we focus on the improvement of the accuracy of cuff-based indirect SBP estimation. We propose an algorithm for the detection of the time instant when cuff pressure (CP) is equal to the maximum arterial pressure (supposed to be equal to SBP) both during inflation and during deflation of the cuff using photoplethysmography (PPG) and the ECG signal. Results of the application of the algorithm for recordings of healthy senior adults and healthy young adults are also reported. The paper does not deal with the application of PPG in cuffless estimation of blood pressure, where the research aim is different.

2 MATERIALS AND METHODS

2.1 SBP Estimation Aided by the PPG Signal

PPG is an optical technique for the measurement of cardiac-induced pulsatile changes in tissue blood volume (Elgendi, 2012). Monitoring PPG during BP measurement helps estimate SBP. If the PPG sensor is placed on a finger of the same arm as the cuff is wrapped, the occlusion of the brachial artery by the cuff influences the PPG waveform. When CP exceeds SBP, the brachial artery is completely occluded during the whole heart cycle, and the pulsation in the PPG signal disappears. During deflation, the pulsation in the PPG signal reappears when CP falls below SBP. Accurate designation of the time instants corresponding to the disappearance and reappearance of PPG pulsation is not straightforward, especially if the signal-to-noise ratio is unfavorable. In some studies, the authors designated these disappearance and reappearance instants by visual inspection (Jönsson, Laurent, Skau & Lindberg, 2005; Nitzan et al., 2005; Nitzan et al., 2013). Visual inspection may be appropriate for research purposes, but for devices used in healthcare, automated methods are required. Lubin et al. considered PPG pulses absent if the value of the AC signal amplitude was lower than 20 % of its maximum value (Lubin, Vray & Bonnet, 2020). Nitzan et al. investigated both the disappearance and reappearance of PPG pulsation. They considered PPG pulses to disappear if the value of the maximal derivative was lower than 1% of the mean initial maximal derivative. For the detection of the reappearance of the pulsation, the PPG curve was divided into time segments corresponding to heart cycles. Two parameters were calculated in each segment: an area parameter, which is related to the pulse waveform and the cross-correlation of the signal in each segment with the signal in the neighboring segments. The authors considered PPG pulses to reappear if the calculated values of the area and the cross-correlation parameters met certain detection criteria. The authors measured PPG on the two index fingers and used both signals for the calculation of parameters (Nitzan, Patron, Glik & Weiss, 2009). We have found that the 1% threshold value of the maximal derivative could be applied to the recordings only after high-order filtering of the PPG signal. However, high-order filtering distorts the signal, which is a potential source of error especially when the derivative of the signal is small. Therefore, we did not utilize the derivative of the PPG signal to detect the disappearance of pulsation. Our proposed method is new in incorporating both the amplitude information in AC signal and the correlation between subsequent heart cycles of the PPG curve, and in utilizing the DC component of the PPG signal. The ECG signal recorded in parallel with PPG gives valuable information when the PPG amplitude is small.

2.2 New SBP Estimation Method Aided by the DC-coupled PPG Signal and the ECG Signal

Our proposed method designates the disappearance and reappearance of the pulsation in the PPG signal by applying similar operations independently both during inflation and deflation. The method is based on parallel recording of three signals, the CP of the cuff wrapped around the upper left arm, the PPG measured on the left index fingertip and the ECG in Einthoven II-lead. The algorithm consists of three stages.

At the beginning of the first stage, the PPG signal is bandpass filtered between 0.5-8 Hz and inverted so that upward signal corresponds to higher blood volume in the fingertip. After that, the difference between the maximum and minimum points of the PPG signal (AC amplitude) is calculated in a sliding window. The window length was empirically determined and set to 500 ms. This window length is long enough to include the whole systolic upstroke segment of the PPG curve even in case of low heart rate. Next, the algorithm locates the intervals both for inflation (Intinfl) and for deflation (Intdefl) where the sliding window AC amplitude is permanently below 10 percent of its maximum value.

In the final step of the first stage of the algorithm, the point is searched for where the sliding window AC amplitude falls below a threshold level. The threshold levels are calculated as fractions of the average AC amplitude over Intinfl and Intdefl. As a result, a short interval is allotted both for inflation and for deflation, where the CP equals the maximum BP.

In the second stage of the algorithm, the first local maximum and local minimum are searched for in the bandpass filtered PPG signal that are at the beginning of the search intervals for inflation and deflation. These points are designated by t_{PPGmax_infl}, t_{PPGmin_infl} (and t_{PPGmax_defl}, t_{PPGmin_defl} for deflation). Two more local maxima are localized preceding t_{PPGmax_infl} (in case of deflation following t_{PPGmax_defl}), in order to delineate two adjacent heart cycles that are required for correlation calculation. Heart cycles are delineated based on local maxima; three successive local maxima enable the delineation of two heart cycles. For the localization of PPG local maxima and minima, the ECG signal is also used, as minima and maxima are searched for within the time interval between corresponding neighboring R-peaks. After the identification of heart cycles, the correlation coefficient (CC) is calculated between the two adjacent heart cycles in the PPG signal, similarly to (Nitzan et al., 2009). The shorter heart cycle is stretched by interpolation in order to have two heart cycles of equal length. Besides the correlation coefficient, PWTT is calculated between the corresponding R-peak in the ECG signal and t_{PPGmin_infl} (t_{PPGmin_defl} for deflation). If CC is larger than 0.85 and PWTT is between 100-500 ms, then a valid PPG pulse belongs to t_{PPGmax_infl} and t_{PPGmin_infl} (or t_{PPGmax_defl} and t_{PPGmin_defl} in case of deflation). Otherwise the supposed two adjacent heart cycles do not exist, the pattern is considered to be noise. In case of a valid PPG pulse, t_{PPGmax_infl} and t_{PPGmin_infl} are moved to the corresponding points in the following heart cycle; CC and PWTT are also recalculated. In case of noise, t_{PPGmax_infl} and t_{PPGmin_infl} are moved to the corresponding points in the preceding heart cycle; CC and PWTT are also recalculated. The procedure continues until the first heart cycle when change in PPG is considered noise in the direction of following heart cycles, or valid pulse in the direction of preceding heart cycles. If one heart cycle is considered as noise, the algorithm examines the following two heart cycles and the preceding heart cycle, calculates CC between all possible pairings of heart cycles and takes the maximum of these CC values. This step is necessary to identify irregular or noisy heart cycles (at most two consecutive heart cycles) that are followed by at least one valid pulse. The outputs of the second stage $t_{stage2\_infl\_max}$ and $t_{stage2\_infl\_min}$ are the time instants when the PPG signal has local maximum and local minimum during the last heart cycle that results in pulsation. In case of deflation, the direction of moving t_{PPGmax_defl} and t_{PPGmin_defl} is inverted compared to inflation, the first heart cycle resulting in pulsation is searched for and the outputs of the second stage are called $t_{stage2\_defl\_max}$ and $t_{stage2\_defl\_min}$.

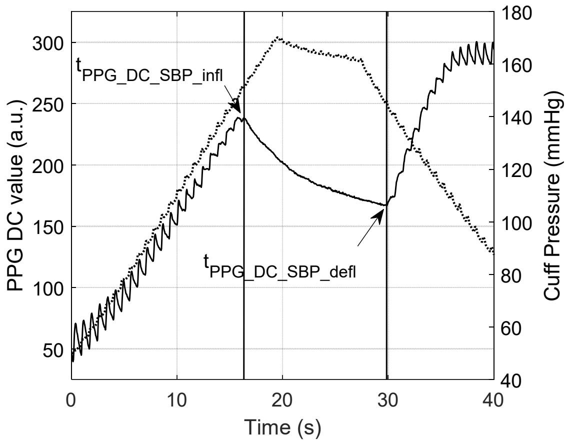

Figure 1: Typical DC-coupled PPG signal (solid line) and the corresponding CP signal (dotted line). Left vertical line: estimated time instant where CP exceeds SBP during inflation. Right vertical line: estimated time instant where CP falls below SBP during deflation.

In the third stage of the algorithm, DC level of the PPG signal is also considered. Instead of the bandpass filtered PPG signal, the lowpass filtered signal is used; the cutoff-frequency of the filtering is 8 Hz. A typical DC-coupled PPG signal and the CP signal recorded in parallel are plotted in Fig. 1.

There is an upward trend in the DC level during inflation and then the DC level starts falling when CP is close to SBP during inflation (left vertical line). The falling trend ends, and the DC level starts rising again when CP is close to SBP during deflation (right vertical line). The algorithm detects the points in the DC-coupled PPG signal, where the trend changes as possible candidates of time instants when CP equals SBP during inflation and deflation. These points are denoted by $t_{PPG\_DC\_SBP\_infl}$ and $t_{PPG\_DC\_SBP\_defl}$. Then, $t_{stage2\_infl\_max}$ is compared to $t_{PPG\_DC\_SBP\_infl}$ and $t_{stage2\_defl\_min}$ is compared to $t_{PPG\_DC\_SBP\_defl}$, because the trend change corresponds to a local maximum during inflation and a local minimum during deflation. If $t_{stage2\_infl\_max}$ is located following $t_{PPG\_DC\_SBP\_infl}$, then $t_{PPG\_DC\_SBP\_infl}$ is rejected because a valid PPG pulse was found following it. In that case, $t_{stage2\_infl\_max}$ and $t_{stage2\_infl\_min}$ are considered and the time instant of the next local minimum in the bandpass filtered PPG signal is extrapolated. This time instant is denoted by t_{xtrp_infl}. The extrapolation is based on the length of the previous heart cycle. The time instant where CP exceeds SBP is estimated as the midpoint between $t_{stage2\_infl\_max}$ and t_{xtrp_infl}. If $t_{stage2\_infl\_max}$ is located preceding $t_{PPG\_DC\_SBP\_infl}$, then a distinction is made depending on whether the distance of the two points is larger than the corresponding heart period (the length of the heart cycle) or not. If the distance is larger, then $t_{PPG\_DC\_SBP\_infl}$ is rejected because it is more than one heart cycle away from the last found valid PPG local maximum. In that case, $t_{stage2\_infl\_max}$ and $t_{stage2\_infl\_min}$ are considered, t_{xtrp_infl} is designated and the time instant where CP exceeds SBP is estimated as the midpoint between $t_{stage2\_infl\_max}$ and t_{xtrp_infl}. If the distance between $t_{stage2\_infl\_max}$ and $t_{PPG\_DC\_SBP\_infl}$ is smaller than one heart period, then $t_{PPG\_DC\_SBP\_infl}$ is accepted as the estimate of the time instant where CP exceeds SBP. In case of deflation, if $t_{stage2\_defl\_min}$ is located preceding $t_{PPG\_DC\_SBP\_defl}$, or the distance between the two time instants is larger than one heart period, then extrapolation is done similarly to inflation, but in the opposite direction. The time instant where CP falls below SBP is estimated as the midpoint between the extrapolated local maximum in the bandpass filtered PPG signal and $t_{stage2\_defl\_min}$. If $t_{stage2\_defl\_min}$ is located following $t_{PPG\_DC\_SBP\_defl}$, and the distance between the two time instants is smaller than one heart period, then $t_{PPG\_DC\_SBP\_defl}$ is accepted as the estimate of the time instant where CP falls below SBP. Characteristic points designated by the algorithm in the bandpass filtered PPG signal are illustrated in Fig. 2.

Our proposed method aims to estimate SBP with a temporal resolution better than one heart cycle. This means that not only heart cycles in the PPG signal are categorized as valid PPG pulse or noise, but also the moment when CP equals SBP is searched for within one heart cycle. If the temporal resolution of SBP detection is only one heart cycle, the error of the estimated SBP difference between inflation and deflation can be high, as demonstrated in Fig. 3.

In the given example, if it is not known, where the time instant corresponding to SBP is located within one heart cycle, the estimated SBP difference between inflation and deflation can be between 5-15 mmHg.

2.3 Data Acquisition

The proposed method was tested on recordings taken by a home health monitoring device (Nagy & Jobbágy, 2018). The device contains an inflatable cuff with two control valves, and it is able to keep cuff pressure at a constant value. The device also measures ECG in Einthoven II-lead and PPG at the fingertip with a sampling frequency of 1 kHz. A transmission-type PPG sensor is used to reduce the effect of motion artifacts. The device contains no hardware filters for ECG and PPG, in order to record undistorted signals. As a result, DC-coupled PPG signal can be measured. The device inflates the cuff with approximately 6 mmHg/s speed to 170 mmHg, then deflates the cuff with the same speed. Deflation is stopped at 60 mmHg for 10 seconds. When 40 mmHg is reached during deflation, CP changes abruptly to 0 mmHg. (The CP=constant section makes possible the analysis and compensation of breathing. Our results in this field are not reported in this paper.)

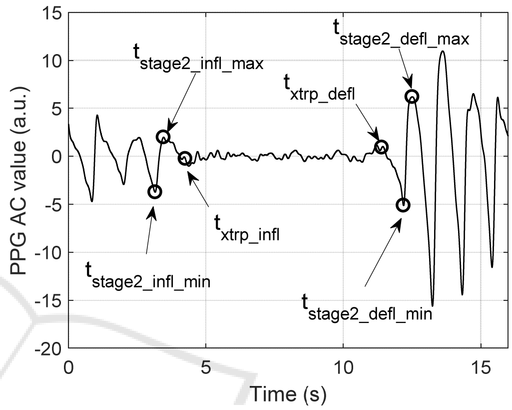

Figure 2: Illustration of characteristic points designated by the algorithm in the bandpass filtered PPG signal.

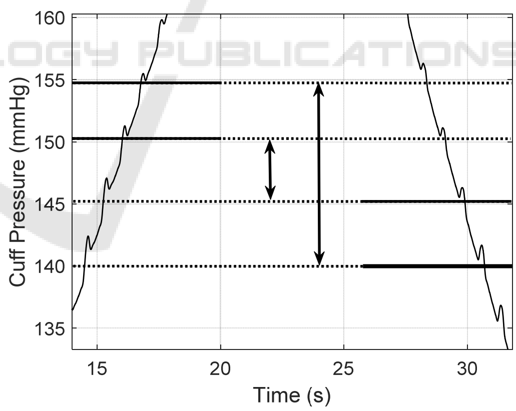

Figure 3: CP signal during inflation and deflation close to SBP. If the temporal resolution of SBP detection is one heart cycle, the error of the estimated SBP difference between inflation and deflation can be high.

2.4 Tested Persons

5 healthy senior persons (age 55-64 years, 2 females, 3 males) and 13 healthy young persons (age 19-35 years, 6 females, 7 males) participated in the measurement series reported in this paper. All tested persons gave their informed consent. The research

was performed in accordance with the Declaration of Helsinki and the study protocol was approved by the Scientific and Research Committee of the Hungarian Medical Research Council (SE RKEB 46/2020).

2.5 Measurement Protocol

In order to validate the proposed new algorithm, three measurement series were used.

SBP values calculated by the algorithm are denoted by SBP_{infl} during inflation and SBP_{defl} during deflation. We denote the difference between SBP_{infl} and SBP_{defl} by ΔSBP as defined in (1).

$$\Delta SBP = SBP_{infl} - SBP_{defl} \tag{1}$$

The first measurement series assessed the effect of physical stress on ΔSBP. Each person performed 20 squat jumps. 8 healthy young adults participated in the measurement series. For each person, one measurement was recorded by the home health monitoring device before the exercise and immediately after the exercise, in sitting position.

The second test series compared the ΔSBP between seniors and young adults, measurements of 5 healthy seniors and 5 healthy young adults were analyzed. All measurements were recorded in resting state of the tested person, in sitting position. 3 measurements were recorded for each person.

The third measurement series aimed at testing the within-subject variability of ΔSBP, measurements of 4 healthy young adults were analyzed. All measurements were recorded in resting state of the tested person, in sitting position. 8 measurements were recorded for each person.

3 RESULTS

Table 1 – 3 show results for the effect of physical stress, ΔSBP difference between seniors and young adults and within-subject variability, respectively. The proposed algorithm gives a good estimate of SBP_{infl} and SBP_{defl}. The reference SBP_{infl} and SBP_{defl} values (used as gold standards) were determined by visual evaluation: the CP value when pulsation in PPG disappears or reappears. Based on the ECG R peak position we could restrict the range within every heartbeat where pulsation in PPG can be present. This helps especially when the signal-to-noise ratio is low. 50 recordings were selected for visual analysis to evaluate the accuracy of the algorithm including 5 recordings after physical exercise, 15 recordings from healthy seniors and 30 recordings from healthy young subjects.

Table 1: The effect of physical stress.

Person Identifier[a]	ΔSBP (mmHg)		
	Before exercise	After exercise	Change
Y1	4.4	-2.8	-7.2
Y2	7.1	5.6	-1.5
Y3	-4.2	9.3	13.5
Y4	3.7	-1.3	-5.0
Y5	11.9	12.7	0.8
Y6	1.7	-0.8	-2.5
Y7	7.1	4.8	-2.3
Y8	-3.8	4.3	8.1
Y1	4.4	-2.8	-7.2
Y2	7.1	5.6	-1.5

a. Eight healthy young subjects.

Table 2: Senior-young group average.

Group[a]	ΔSBP (mmHg)		
	Mean[b]	Min	Max
Senior	-1.2	-11.5	10.6
Young	4.1	-5.0	11.6

a. Five senior healthy subjects, five young healthy subjects.
b. Three measurements taken for each subject, sitting position, resting state.

Table 3: Within-subject variability.

Person Identifier[a]	ΔSBP (mmHg)		
	Mean[b]	Min	Max
Y9	5.1	-0.1	8.2
Y10	5.8	-5.0	14.2
Y11	8.3	-0.1	13.4
Y12	7.0	-3.3	15.2

a. Four young healthy subjects.
b. Eight measurements taken for each subject, sitting position, resting state.

4 DISCUSSION

The values determined by the suggested new algorithm are within a narrow range around the gold standard. The mean value and standard deviation of the difference between values determined by the algorithm and the gold standard was -0.7 ± 1.5 mmHg for SBP_{infl} and -0.3 ± 1.8 mmHg for SBP_{defl}. In 88 % of the 50 examined recordings, the absolute error of the algorithm compared to the gold standard is below 3 mmHg. Considering the 6 mmHg/s speed of inflation and deflation, this corresponds to a temporal resolution much better than one heart cycle. The proposed new algorithm assures better than usual resolution in SBP determination. This is necessary for the appropriate evaluation of the three measurement series.

CP values corresponding to time instants when CP is supposed to be equal to SBP indicated by the PPG signal may considerably differ during inflation and deflation. Both for seniors and for young adults, the largest absolute difference found was more than 10 mmHg. This difference cannot be evaluated using the usual indirect BP measurement methods. The method proposed in this paper assures the necessary resolution and accuracy.

The effect of physical stress on ΔSBP showed large variability among individuals, in both magnitude and direction. The change in ΔSBP was negative for 5 persons and positive for 3 persons, with the smallest absolute value of 0.8 mmHg and the largest absolute value of 13.5 mmHg. Sign of ΔSBP changed for five persons because of exercise. Interestingly, the person showing the largest ΔSBP value before exercise (Y5) showed the smallest change in ΔSBP because of exercise. ΔSBP gives information basically on the circulation in the upper arm.

Group-level comparison of senior and young adults revealed negative mean ΔSBP value for seniors and positive mean ΔSBP value for young adults. Although negative ΔSBP values appeared also for young individuals, this result suggests that people with increased arterial stiffness tend to produce more negative ΔSBP values.

ΔSBP values showed high within-subject variability (more than 8 mmHg difference of the minimum and maximum values) for all four tested persons although the mean ΔSBP was higher than 5 mmHg for each of them. This result shows that one single measurement of ΔSBP may not be sufficient to estimate the state of the cardiovascular system of a tested person.

DC level of the PPG signal contains valuable information aiding SBP estimation. However, valid pulses were found following $t_{PPG\_DC\_SBP\_infl}$ and preceding $t_{PPG\_DC\_SBP\_defl}$ in some cases suggesting that the DC level of PPG is influenced by more factors. Occlusion of the artery by the cuff is a dominant factor as DC level shows a rising trend during inflation in all cases.

We compared our results with the work of Zheng et al. who investigated the mechanical behavior of the brachial artery during cuff inflation and deflation (Zheng, Pan & Murray, 2013). The authors compared SBP, DBP and MAP values measured during inflation and deflation of the cuff by manual auscultatory and automated oscillometric methods. SBP from inflation was found to be statistically significantly lower than SBP from deflation, measured by both the manual and the automated method. This result was not justified by our measurements. The difference in SBP during inflation and deflation was studied also by other researchers (Fabian et al., 2016).

SBP_{infl} and SBP_{defl} are close to each other. Comparing them needs a measurement method with good resolution. In this regard, the method we propose outperforms other methods. It finds the disappearance and reappearance of the PPG signal with better time resolution than a heartbeat while other methods can detect these fiducial points with heartbeat resolution only.

5 CONCLUSIONS

Accurate indirect BP measurement is the cornerstone of the detection and management of hypertension. In this paper, we proposed a new method for the estimation of SBP using CP, ECG and DC-coupled PPG signals. It eliminates the most severe device dependent limitation associated with indirect BP measurement, thus our method offers more accurate non-invasive estimation than the presently available other methods even in the case of arrhythmia. In the near future, the algorithm will be applied to patients with Left Ventricular Assist Device, LVAD. The non-invasive blood pressure measurement of these patients is important but cumbersome, requiring special expertise. Our proposed method is applicable for them also at home, without medical expert.

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