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# Radial Electrical Impedance: A Potential Indicator for Noninvasive Cuffless Blood Pressure Measurement

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## Abstract

Noninvasive, cuffless, and continuous blood pressure (BP) monitoring is essential to prevent and control hypertension. A well-known existing method for this measurement is pulse transit time (PTT), which has been investigated by many researchers as a promising approach. However, the fundamental principle of the PTT method is based on the time interval taken by a pulse wave to propagate between the proximal and distal arterial sites. Consequently, this method needs an independent system with two devices placed at two different sites, which is a problem. Even though some studies attempted to synchronize the system, it is bulky and inconvenient by contemporary standards. To find a more sensitive method to be used in a BP measurement device, this study used radial electrical bioimpedance (REB) as a potential indicator for BP determination. Only one impedance plethysmography channel at the wrist is performed for demonstrating a ubiquitous BP wearable device. The experiment was evaluated on eight healthy subjects with the ambulatory BP monitor on the upper arm as a reference. The results demonstrated the potential of the proposed method by the correlation of estimated systolic (SBP) and diastolic (DBP) BP against the reference at 0.84 ± 0.05 and 0.83 ± 0.05, respectively. REB also tracked the DBP well with a root-mean-squared-error of 7.5±1.35 mmHg.

**Keywords:** Blood pressure, Radial artery, Electrical impedance, Plethysmography

## 1. INTRODUCTION

Hypertension is a critical medical condition and a risk factor for several other diseases. According to [1], the number of hypertensive patients is anticipated to increase by 2025. Therefore, hypertension should be treated and controlled. To monitor this disease, the patient's blood pressure (BP) is measured regularly, especially in elderly hypertensive people. Hence, the BP measurement method should not be invasive, interruptive, or manual. Several non-invasive, cuffless BP methods are now available, such as volume clamping and tonometry. At the present time, the most promising method, which meets all the above requirements, is the pulse transit time (PTT) method.

Many studies have demonstrated the cuffless continuous BP system that is based on PTT method [2-4]. PTT is the time taken for the propagation of blood between the proximal and distal arterial sites. The fundamental principle of PTT method is that high BP corresponds to a short PTT value and vice versa [5]. When the BP increases, the pulse wave velocity (PWV) is found to be faster than usual. Correspondingly, PTT value, which is reversely proportional to PWV, will decrease. The relationship between BP and PTT is given by:

$$PWV = \frac{D}{PTT} = \sqrt{\frac{E_0 e^{\alpha P} h}{\rho d}} \quad (1)$$

where $E_0$ and $\alpha$ are subject-specific parameters, $\rho$ is the blood density, $h$ and $d$ are the arterial properties, and $D$ is the distance between two arterial sites [6]. Even though PTT is a promising approach for cuffless continuous BP measurement, there are several problems yet to be solved. First, it is easy to recognize that the PTT method requires two pulse waveforms to calculate the transit time between two measured sites. If the distance $D$ is too far, it would not be convenient for the patient and for long-term BP recording. For the proximal waveform, typically, the electrocardiography (ECG) is used in the chest and is the most well-known method. To detect the waveform in the peripheral artery such as the finger, toe, or ear,

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photoplethysmography (PPG) is performed on the distal waveform [3]. Consequently, PTT can be computed from the R-wave of ECG to the PPG foot point in the same cardiac cycle. To minimize the power consumption by a 24-hour BP measurement device, wireless communication, which can enable two separate devices to communicate, is not applied. To solve this problem, some studies reduced the distance between the two measured placements to increase convenience [7,8]. Based on the analog properties of the pulse waveform, however, a more convenient system achieves less accuracy [9]. Therefore, there are limitations with the PTT method.

The second major problem of the PTT method is the lack of accuracy. With the well-known PTT model [10], as shown in (2) and (3), the systolic BP (SBP) and diastolic BP (DBP) are estimated using two independent determinations.

$$SBP = K_1 \cdot \ln(PTT) + K_2 \tag{2}$$

$$DBP = K_3 \cdot \ln(PTT) + K_4 \tag{3}$$

Here, $K_i$ are unknown subject-specific parameters. It is obvious that a single PTT might not estimate two BP values as indicated by certain previous studies [10-12]. This was confirmed in [2], where, to improve both the BP determinations, the PPG intensity ratio was applied in combination with PTT method.

One potential solution that can solve the aforementioned problems of PTT method is arterial impedance, which is directly reflected by the BP. The arterial impedance has been reported as an indirect indicator related to the BP as shown in [13, 14]. As shown in Fig. 1, this study investigates a novel BP algorithm using only one radial electrical bioimpedance (REB) channel to estimate both BP values. The proposed system was performed at the radial artery, which is very close to the skin surface, for demonstrating the convenience in the new approach of noninvasive cuffless BP monitoring.

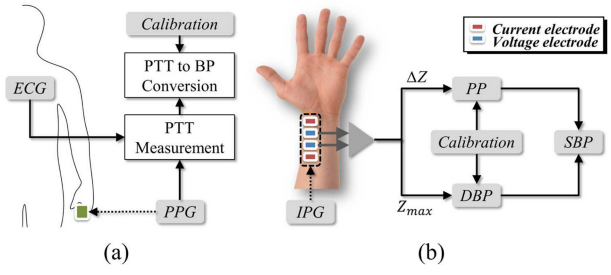


**Fig. 1.** (a) Typical PPT method using ECG and PPG: high BP value corresponds to short PTT value and vice versa. (b) REB measurement at wrist: Radial impedance is affected by blood volume.

## 2. METHODOLOGY AND SYSTEM

### 2.1 Blood Pressure Estimation

The measured body segment is assumed to be modeled as a cylindrical section, which contains the tissue and the radial artery. Therefore, as shown in (4), the total measured impedance $Z$ is a parallel connection of the arterial impedance $Z_a$ and the tissue impedance $Z_t$ [15]. Owing to the tissue impedance remaining a subject-specific constant, the change in the measured impedance $\Delta Z$ can be directly monitored from the change in arterial impedance.

$$1/Z = 1/Z_a + 1/Z_t \tag{4}$$

When a small blood volume change occurs, $\Delta Z$ is directly proportional to the change in arterial radius $\Delta R$ by excluding the change in the mean arterial radius $R_m$ as follows.

$$\Delta A = 2\pi R_m \Delta R = \rho L \Delta Z / Z_0^2 \tag{5}$$

where $L$ and $Z_0$ represent the length and the original impedance of the measured segment, respectively. On the other hand, the change in BP or pulse pressure (PP) causes a change in the arterial radius. Thus, PP is a function of $\Delta R$ with the elastic modulus $E$, the internal radius $R_i$ and $\sigma$ as Poisson's ratio [16]. Accordingly, PP is in direct proportion to $\Delta R$ as well as $\Delta Z$ on the assumption that the elastic modulus $E$ remains constant under consideration.

$$E = \frac{PP}{\Delta R}\frac{2(1-\sigma^2)R_0 R_i^2}{R_0^2 - R_i^2} \tag{6}$$

$$PP \propto \Delta R \propto \Delta Z \tag{7}$$

For DBP estimation, the systemic arterial system is modeled as an electrical equivalent circuit consisting of a capacitor (to represent the arterial compliance $C$) and a resistor (to represent the total peripheral resistance $TPR$). Hence, the resulting DBP is given by:

$$DBP = P_0 \cdot e^{\frac{-T}{TPR \cdot C}} \tag{8}$$

where $P_0$ is the end-systolic aortic pressure and $T$ is the heart period [17]. On the other hand, $TPR$ is directly proportional to the second power of the arterial impedance as well as the total measured impedance through the arterial cross-sectional area as shown below [18].

$$TPR = \frac{8\pi L\eta}{A^2} = \frac{8\pi L\eta}{(\rho L / Z_a)^2} \propto Z^2 \tag{9}$$

Thus, according to (8) and (9), under some assumptions in a relatively short period, DBP is a function of $Z$ and the heart period $T$ in which $K$ is a subject-specific constant.

$$DBP \propto e^{-K\frac{T}{Z^2}} \tag{10}$$

Based on (7) and (10), with some initially calibrated values ($PP_0$, $\Delta Z_0$, $DBP_0$, $K$, $Z_0$, $T_0$), the BP values can be derived.

$$PP = PP_0 \frac{\Delta Z}{\Delta Z_0} \tag{11}$$

$$DBP = DBP_0 e^{K\left(\frac{T_0}{Z_0^2}-\frac{T}{Z^2}\right)} \tag{12}$$

$$SBP = DBP + PP \tag{13}$$

## 2.2 Radial Impedance Measurement

For radial impedance measurement, impedance plethysmography measurement is the most popular noninvasive method that has been widely used. The fundamental principle of the REB method is that a constant alternating current source is injected into the skin surface, following which the measured voltage can reflect the impedance segment according to Ohm's law [15]. The injected current source is 1 mA at 100 kHz which is strong enough to propagate through the skin while meeting the safety standard requirements for medical electrical equipment [19]. The proposed system is built upon tetrapolar measurement consisting of two pairs of electrodes for current injection and voltage measurement. For the electrode material, metal copper foils were experimented to demonstrate the potential of low-cost dry-contact electrodes.

As shown in Fig. 2, the first stage of the REB system is a high-speed instrumentation amplifier (IA) AD8220, which maximizes dynamic range on the low voltage supplies in portable applications. After using the amplitude demodulation for the output of IA, the 100 kHz carrier signal is removed whereby the impedance waveform is achieved including the basal impedance $Z_{basal}$ and the impedance variation $\Delta Z$. Thereupon, the high pass filter is applied to eliminate $Z_{basal}$ with a cutoff frequency of 0.05 Hz. Further, to get a clear radial impedance waveform without undesired noises such as motion artifact, the low pass filter response is performed with the selected corner frequency as 3 Hz. Finally, the received signal will be amplified for digital signal processing.

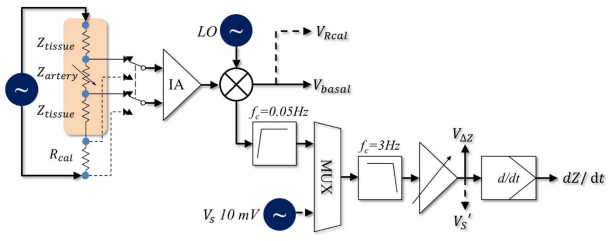


**Fig. 2.** The proposed system for radial electrical bio-impedance measurement.

Using the known resistor $R_{cal}$ and voltage source $V_S$ at 10 mV, the actual radial impedance values can be calibrated with $V_{Rcal}$ as the voltage drop over $R_{cal}$, $V_{basal}$ as the output voltage of demodulation stage and $V_S'$ as the measured voltage after final amplification corresponding to $V_S$.

$$Z = Z_{basal} + \Delta Z = \frac{R_{cal}}{V_{Rcal}}\left(V_{basal} + \frac{V_S}{V_S'}V_{\Delta Z}\right) \tag{14}$$

To minimize the sensitivity of heart rate detection, the differentiator is applied. The heart period T is determined by the period between two consecutive peaks of IPG's first derivative.

## 2.3 Validation Study

In this study, ambulatory BP monitor Omron HEM-790IT, which meets three internationally recognized standards (AAMI [20], BHS [21], ESH [22]), is used to obtain the reference BP values. Eight healthy adults (five males, three females) ages 30 ± 6 years, without any history of conditions such as hypertension and cardiovascular conditions, were enrolled. The subjects were instructed to relax for 10−15 minutes in the seated position before verifying BP accuracy. REB sensor and the reference device were placed on the subjects' left arm and held over the chest area during the recording time as shown in Fig. 3(a). To get a variety of BP values, hemodynamic intervention was designed. First, the subject was made to relax to obtain the initial BP value and was then made to perform a handgrip exercise for 2 min to increase BP values. For each subject, six measurements were recorded from three baseline recordings and three interventions, alternately. The mean of all beats over a 20 s period from the proposed system was used as one data point. The BP value from the reference was obtained later. Initially, the proposed system was calibrated, following which six pairs of BP values were evaluated. To assess the proposed system, the correlation coefficient (r) and the root-mean-squared-error (RMSE) between the estimated and reference BP values per subject were computed.

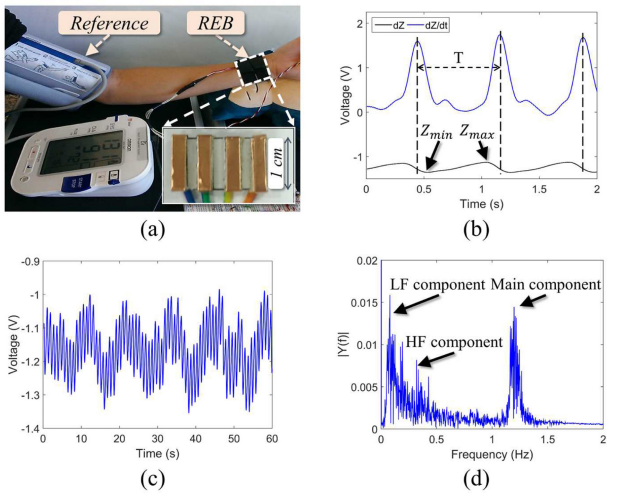


**Fig. 3.** (a) Radial impedance is measured at wrist, whereas reference device is located in the upper arm, (b) Heart period and impedance parameters calculation, (c) Continuous radial impedance waveform, (d) The frequency spectrum of REB signal.

## 3. RESULTS

### 3.1 Radial Impedance Analysis

The radial impedance waveform and its *dZ/dt* waveform of a representative subject are shown in Fig. 3(b). Corresponding to each *dZ/dt* peak, the maximum $Z_{max}$ and minimum $Z_{min}$ radial impedance were detected and the impedance variation was calculated. Since the arterial cross-sectional area is directly proportional to BP [23] and is in inverse proportion to arterial impedance in (15), the higher value of BP corresponds to the lower value of radial impedance and vice versa. Hence, $Z_{max}$ can be used to estimate DBP value.

$$A = \rho L / Z_a \propto 1 / Z \tag{15}$$

Typically, a continuous BP signal has a main component which is caused by the heart activity and corresponds to the heart rate (around 1 Hz). Moreover, arterial pressure also contains two major components, namely, a high-frequency (HF) component (fast variability) at around 0.3 Hz and a low-frequency (LF) component (slow variability) at around 0.1 Hz. As seen in Fig. 3(d), in the frequency spectrum of the radial impedance waveform, the HF and LF variations of BP can be found at the same frequency with those of the REB signal. These obvious results demonstrate that radial impedance can be used to determine both SBP and DBP.

### 3.2 Blood Pressure Validation

The correlation coefficients and RMSEs of predicted SBP and DBP against with reference device are summarized in Table 1. On

**Table 1.** Correlation coefficients and RMSEs between the estimated BP and reference device for each subject

| Sub No. | Pearson r | | RMSE (mmHg) | |
|---|---|---|---|---|
| | SBP | DBP | SBP | DBP |
| 1 | 0.85 | 0.88 | 13.07 | 8.29 |
| 2 | 0.89 | 0.8 | 12.3 | 6.12 |
| 3 | 0.78 | 0.79 | 11.41 | 8.72 |
| 4 | 0.88 | 0.9 | 13.96 | 8.64 |
| 5 | 0.77 | 0.82 | 13.29 | 7.31 |
| 6 | 0.81 | 0.79 | 12.56 | 6.79 |
| 7 | 0.91 | 0.9 | 9.43 | 5.26 |
| 8 | 0.85 | 0.78 | 10.67 | 8.86 |
| Mean | 0.84 | 0.83 | 12.09 | 7.5 |
| SD | 0.05 | 0.05 | 1.5 | 1.35 |

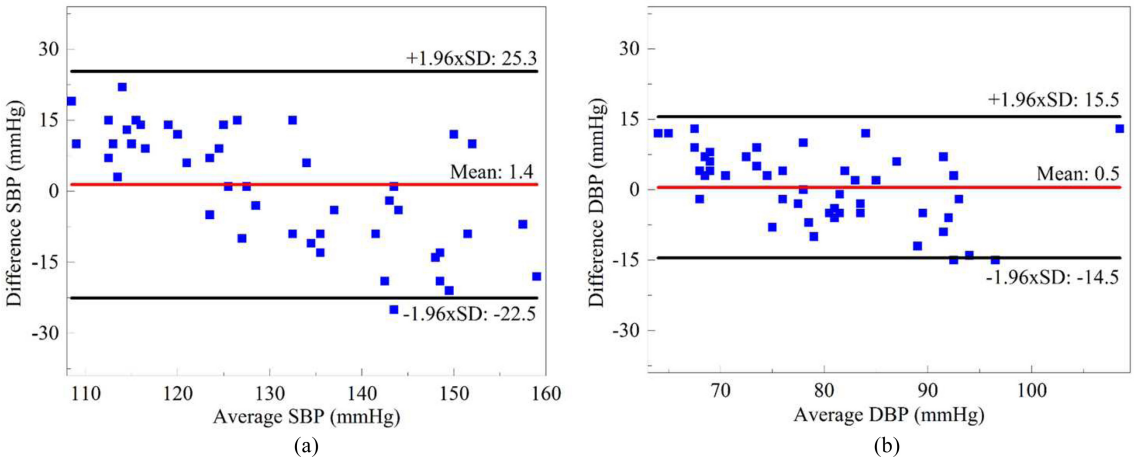


**Fig. 4.** Bland−Altman plots of estimated SBP (a) and DBP (b) against with Omron HEM-790IT, respectively.

an average, the proposed system shows strong correlations for both SBP (0.84 ± 0.05) and DBP (0.83 ± 0.05). These results indicated a strong positive linear relationship between the two devices. Furthermore, the proposed system yields a good RMSE (7.5 ± 1.35 mmHg) for DBP, whereas the SBP RMSE (12.09 ± 1.5 mmHg) shows that the REB system does not track BP changes well. The large SBP changes can explain why the REB is able to access DBP better than SBP. To analyze in detail, the difference and average of SBP and DBP between two systems were obtained. The means ± SD were −1.4 ± 12.2 and 0.5 ± 7.7 mmHg for SBP and DBP, respectively. Based on that, the limits (mean±1.96×SD) were computed as shown in the Bland–Altman plots in Fig. 4.

## 4. DISCUSSION AND CONCLUSION

In this study, radial impedance is used as an indicator to estimate both SBP and DBP. The proposed system is performed at the wrist to demonstrate the potential of a wearable BP measurement device. In comparison with most noninvasive cuffless BP methods (like the PTT method), some major existing problems such as the need for an independent system of two devices and determination of both BP values independently, were resolved as shown in the experimental results. The system for REB measurement was also designed with small electrodes on a small local area. In addition, the relationship between BP and arterial impedance has been reported in detail while some studies just showed this in an indirect form [13, 14]. Apart from the effect of impedance, the heart period is also used for BP determination.

However, there are some limitations yet to be overcome. First, even though the filter is applied to eliminate the effects of motion artifacts, it still affects the impedance amplitude during strenuous activity. Therefore, the proposed system should be used only for elderly people or for overnight measurement. For further applications, this solution must be improved. Second, instead of oscillometry, a mercury sphygmomanometer should be used to validate the standard BP measurement. Also, in the experimental validation, the number of subjects is not large enough to meet the standard protocol and all subjects are normotensive. Finally, the standard location for BP measurement is the upper arm, whereas the proposed system is located at the wrist. However, wrist monitors will have some advantages and are being investigated towards the development of a new BP device.

In summary, the radial impedance shows a strong correlation with BP and has been studied as a potential indicator for BP estimation. In further research, the radial impedance measurement will be developed or combined well with PTT method.

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