Cuffless Continuous Estimation of Relative Mean Arterial Pressure Using Unrestrained and Noncontact Ballistocardiogram and Electrocardiogram: Evaluation in Short Time In-bed Experiments

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Abstract To facilitate capturing the characteristic variations of blood pressure (BP) such as BP surges during the nocturnal period, the in-bed continuous daily measurement of BP may be useful. In this study, we proposed and evaluated a method for cuffless continuous estimation of relative mean arterial pressure (MAP) using capacitive ballistocardiogram (cBCG) and electrocardiogram (cECG) measured using an unrestrained and noncontact method. We adapted a well-known equation for calculating MAP, which is equal to the product of cardiac output and peripheral vascular resistance. We then derived an estimation formula for the relative MAP using the J–K amplitude from the cBCG, heart rate from the cECG, and pulse beat arrival time calculated from the cBCG and cECG. To determine the coefficients for the estimation formula, we measured the MAP of a subject with a commercial device and used the least squares method. To obtain input data for the estimation formula, the cBCG from the heel of the right leg and cECG from the back of the subject were measured simultaneously with capacitively coupled electrodes placed under a bed sheet. The total length of the input data was 80 s for each Valsalva test (VT), and the Valsalva maneuver was used to increase BP during measurement. The data for each VT was separated into a training segment (Tr) and a test segment (Te). To evaluate the proposed estimation method, the following indices were calculated for each VT in 7 subjects: (1) correlation coefficient (CC) between estimated and reference MAP values, (2) confidence interval (CI), and (3) root mean square error (RMSE). For the Tr, average CC was 0.93 ± 0.06, average CI was 2.96 ± 1.29 mmHg, and average RMSE was 0.75 ± 0.33 mmHg. Furthermore, average RMSE for the Te was 2.49 ± 2.22 mmHg. These results indicate that the continuous cuffless method proposed in this study can be used for estimating relative MAP over a short time period. As the subjects in this study were all men in their early twenties, further validation in diverse subjects is required for broad application of the proposed method.

Keywords: cuffless continuous estimation, relative mean blood pressure, ballistocardiogram, electrocardiogram, unrestrained and noncontact method.

Adv Biomed Eng. 10: pp. 36–50, 2021.

1. Introduction

According to a World Health Organization report, cardiovascular disease (CVD) is the main cause of death worldwide [1]. As hypertension is one of the most important modifiable risk factors for CVD [2], early detection and continuous management of hypertension are necessary. Hypertension is classified into three categories: sustained hypertension, masked hypertension, and white coat hypertension [3, 4]. Masked hypertension is diagnosed when a patient exhibits blood pressure (BP) within the normal range in the clinic but above the normal range at home. Ambulatory BP monitoring (ABPM) is used when masked hypertension is difficult to diagnose using clinic and home BP measurements alone, when these measures are suspicious, or when the BP in the clinic or at home changes significantly [4–7]. Masked hypertension is thought to be as high risk for CVD as sustained hypertension [3, 7]. Therefore, early detection of masked hypertension, antihypertensive treatment, and
BP management are needed. Masked hypertension includes nocturnal hypertension, in which BP increases during sleep [8, 9]. Nocturnal BP better reflects the severity of hypertension than daytime BP [5]. Although nocturnal BP has to be measured during sleep, ABPM is the only technique for measuring BP in this context [5]. ABPM techniques include the Korotkoff microphone method and the oscillometric method, both of which measure BP every 15 to 30 minutes by attaching a cuff to the upper arm and applying pressure [5]. However, the cuff pressurization may cause numbness in the upper arm and result in sleep disturbances, which may increase the systolic BP (SBP) [5]. While ABPM with cuff pressurization is acceptable for a single day, it is not always appropriate for long-term and routine BP measurements (ABPM is not positioned as a routine repetitive examination [5]).

Kario [10] reported the resonance hypothesis of BP variability producing BP surge. In that report, BP variability included seasonal, day-to-day, diurnal, and heart beat-to-beat changes, and it is thought that a large surge is caused by their resonance. A large surge may cause CVD when the height and duration of the surge exceed a certain level. Based on the above, there exists a need for a method that allows continuous monitoring of BP during sleep, with beat-to-beat precision, for a long period of time without using a cuff.

Methods for the estimation of BP with cuffless techniques have been extensively studied. These studies focus on cuffless BP estimation using the time interval obtained from electrocardiogram (ECG), seismocardiogram (SCG), phonocardiogram (PCG), impedance cardiography (ICG), photoplethysmography (PPG), and impedance plethysmography (IPG), among others. Seeberg et al. [11] reported a method for BP estimation using the pulse transit time calculated from three types of signals, ECG, ICG, and PPG. Chen et al. [12] and Zheng et al. [13] used the pulse arrival time or pulse transit time from ECG and PPG. Yang et al. [14] reported a method for BP estimation that took the pre-ejection period (PEP) into consideration by calculating the time interval from SCG and PPG, or acoustic techniques. Huynh et al. [15] reported a method to calculate the time interval between PPG and IPG and estimate BP considering PEP without using ECG. Zhang et al. [16] reported the relationship between BP and the time interval between ECG and PCG, by which the sound generated from the movement of the heart valve could be measured. Watanabe et al. [17] performed BP estimation using only PPG and reported high BP estimation accuracy. In all of these methods, the ECG, SCG, PCG, ICG, PPG, and IPG measurements required contact of electrodes with the subject’s skin; they also required measuring devices such as ICG and PPG sensors, to be attached to the chest, fingers, and arms. When measuring BP during sleep on a daily basis for a long period of time, the use of wearable sensors can cause various problems. For example, this technique may not be applicable to individuals with dementia in a care facility, as the discomfort of wearing the devices may cause the person to remove them. Furthermore, the electrodes used in ECG, SCG, and ICG, which are applied directly to the skin, can cause rashes over time. In addition, these measurement techniques require many disposable items that must be changed frequently, and the risk of infection to medical workers during the changes is high. Therefore, eliminating direct contact with the electrodes, by using a cloth between the electrodes and the skin instead, will improve sanitation, reduce the chances of running out the disposable materials, and reduce the risk of infection in medical workers. Thus, we sought to design a method of using biological signals for BP estimation that can be measured using an unconstrained and nonwearable method.

Signals from ballistocardiogram (BCG) can be used for estimating BP and do not require a wearable device. Martin et al. [18] reported estimation of BP using the time interval between BCG and ECG, or BCG and PPG, for subjects in a standing position. In this case, a wearable sensor was used to measure ECG and PPG, but BCG was measured without the wearable device. Takeuchi et al. [19] reported a method to measure capacitive ECG (cECG) and capacitive BCG (cBCG) from the back of a subject in a noncontact and unrestrained manner. Although their report indicated that detecting cBCG signals in this manner was possible, the relationship between the measurement and BP was not stated. Takeuchi et al. [20] and Sakajiri et al. [21] measured cECG and cBCG using a noncontact, unrestrained, and nonwearable method, and calculated the pulse beat arrival time (PBAT), which is the time interval between cECG and cBCG. In those reports, there was a high negative correlation between PBAT and SBP. Taken together, these results suggest that cECG and cBCG measured by a noncontact, unrestrained and nonwearable method, can be used for BP estimation.

Takeuchi et al. [20] reported high negative correlation coefficients (CCs) between PBAT and SBP, and that three of seven subjects showed low CC (|CC| < 0.7) in one of the two Valsalva tests. Based on their findings, we explored how to obtain high correlation between PBAT and BP even for the subjects showing low CC [20], using parameters obtained by the same unrestrained, noncontact method.

2. Materials and Methods

2.1 Estimation of mean arterial pressure

We focus on a well-known equation for determining
mean arterial pressure (MAP), which is equal to the product of cardiac output (CO) and peripheral vascular resistance (PVR) [22]. We modified this formula for estimating relative MAP using cECG and cBCG.

When CO ($F_{CO}$ [L/min]) and PVR ($R_{PVR}$ [dyn·s·cm$^{-5}$]) are given, the value of MAP ($P_{MAP}$ [mmHg]) is expressed as follows [22]:

$$P_{MAP} = F_{CO} \cdot R_{PVR}$$  \hspace{1cm} (1)

$F_{CO}$ is determined using the stroke volume ($V_{SV}$ [L/beat]) and the heat rate ($N_{HR}$ [bpm]) as shown in equation (2) [22, 23]:

$$F_{CO} = V_{SV} \cdot N_{HR}$$  \hspace{1cm} (2)

By substituting Equation (2) into Equation (1), $P_{MAP}$ is expressed as in Equation (3):

$$P_{MAP} = V_{SV} \cdot N_{HR} \cdot R_{PVR}$$  \hspace{1cm} (3)

Recently, He et al. [24] have reported that $V_{SV}$ is proportional to the amplitude of BCG ($V_{BCG}$). Given that, securing the redundancy, $V_{SV}$ is proportional to $V_{BCG}$ raised to the power of $a$ ($V_{BCG}^a$) with a factor $k_1$ as follows:

$$V_{SV} = k_1 \cdot V_{BCG}^a$$  \hspace{1cm} (4)

then, $P_{MAP}$ is expressed as shown in Equation (5):

$$P_{MAP} = k_1 \cdot V_{BCG}^a \cdot N_{HR} \cdot R_{PVR}$$  \hspace{1cm} (5)

In Equation (5), $R_{PVR}$ is known to be proportional to viscosity [25], and the viscosity is considered to decrease with flow velocity ($U$ [m/s]) [26]. Furthermore, with relation to peripheral arterial compliance, it is expected that as $R_{PVR}$ increases, PBAT at peripheral measuring site ($T_{PBAT}$ [ms]) decreases. Therefore, $R_{PVR}$ is assumed to be inversely proportional to both $U$ raised to the power of $b$ ($U^b$) and $T_{PBAT}$ raised to the power of $c$ ($T_{PBAT}^c$), and can be expressed as in Equation (6) using a factor $k_2$:

$$R_{PVR} = \frac{k_2}{U^b \cdot T_{PBAT}^c}$$  \hspace{1cm} (6)

In Equation (6), $U$ is obtained from $F_{CO}$ by changing the unit from [L/min] to [m$^3$/s], and by dividing the unit-converted $F_{CO}$ by the net cross-sectional area (A [m$^2$]) of the flow path as follows:

$$U = \frac{F_{CO}}{60} \cdot \frac{1}{A}$$  \hspace{1cm} (7)

From Equations (2), (4) and (7), $U$ is finally expressed as in Equation (8):

$$U = V_{SV} \cdot N_{HR} \cdot \frac{10^{-3}}{60} \cdot \frac{1}{A}$$

where $k_3 = k_1/(6 \times 10^4 \cdot A)$. By substituting Equation (8) into Equation (6), we obtain $R_{PVR}$ as follows:

$$R_{PVR} = \frac{k_2}{(k_3 \cdot V_{BCG}^a \cdot N_{HR})^b \cdot T_{PBAT}^c}$$  \hspace{1cm} (9)

By substituting Equation (9) into Equation (5), $P_{MAP}$ is expressed by the following:

$$P_{MAP} = k_1 \cdot V_{BCG}^a \cdot N_{HR} \cdot \frac{1}{k_3^b} \cdot \frac{1}{V_{BCG}^{ab} \cdot N_{HR}^{b-1} \cdot T_{PBAT}^c}$$

$$= \frac{k_1 k_2}{k_3^b} \cdot \frac{1}{V_{BCG}^{(b-1)} \cdot N_{HR}^{(b-1)} \cdot T_{PBAT}^c}$$  \hspace{1cm} (10)

By transforming both members into logarithmic representation, we obtain Equation (11) as follows:

$$\ln P_{MAP} = \ln \frac{1}{V_{BCG}^{(b-1)} \cdot N_{HR}^{(b-1)} \cdot T_{PBAT}^c} \cdot \frac{k_1 k_2}{k_3^b}$$

$$= a(1 - b) \ln V_{BCG} + (1 - b) \ln N_{HR} - c \ln T_{PBAT} + \ln \frac{k_1 k_2}{k_3^b}$$  \hspace{1cm} (11)

Replacing $\ln \frac{k_1 k_2}{k_3^b} = d$, Equation (11) can be rewritten as follows:

$$\ln P_{MAP} = a(1 - b) \ln V_{BCG} + (1 - b) \ln N_{HR} - c \ln T_{PBAT} + d$$  \hspace{1cm} (12)

For example, if four or more sets of simultaneously measured $V_{BCG}, T_{PBAT}, N_{HR}$, and reference MAP ($P_{MAP, ref}$ [mmHg]) are available for each subject, all parameters from $a$ to $d$ can be determined (for example, by least square approach). After determining all the parameters, MAP ($P_{MAP, est}$ [mmHg]) is estimated using Equation (13) and data of simultaneously measured $V_{BCG}, N_{HR}$, and $T_{PBAT}$.

$$P_{MAP, est} = e^{a(1-b) \ln V_{BCG} + (1-b) \ln N_{HR} - c \ln T_{PBAT} + d}$$  \hspace{1cm} (13)

### 2.2 Experimental systems and procedures

All experimental procedures were approved by the Human Life Ethics Committee of Tokyo Denki University. Seven healthy male subjects (age: 21–23 years) provided informed consent prior to participation in our experiments. All subjects were included in the study if they had no prior medical history of cardiopulmonary or cardiovascular disease. The age, gender and physical data of the seven subjects are shown in Table 1. Experimental raw data analyzed in this study is identical with that measured.
sured in the previous report [20]. Subjects #A to #G in Table 1 correspond respectively to subjects #1 to #7 in the previous report [20].

**Figure 1** shows the electrode and sensor layout, and block diagram of the measuring systems for cBCG and cECG. Both cBCG and cECG were measured with the systems reported previously [20], using conductive fabric electrodes placed under the bed sheet, in a noncontact and unrestrained manner. For cBCG measurement, two of the electrodes under the sheet were placed at the positions of the heel and the calf of the right leg to form capacitive coupling. Net capacitance of the two capacitive coupling changes with ballistocardiographic motion of the leg. The change in capacitance was detected as the change in frequency with an astable multivibrator. The change in frequency was converted to a change in voltage using a frequency-to-voltage converter. For cECG measurement, two double-shield rectangular electrodes under the bedsheet were arranged in parallel above and below the position of the subject’s heart. Each of the electrode forms capacitive coupling with the back and detects an alternating current (AC) component of the ECG signal via the coupling. Recording of the cBCG and cECG signals was performed using a commercially available analog-to-digital converter (MP150, BIOPAC Systems) and commercially available software (AcqKnowledge4.1, BIOPAC Systems), at 16 bit and sampling frequency of 1.0 kHz.

Relative MAP was estimated by applying the proposed method to the previously obtained experimental data [20]. Each subject was asked to wear commercial pajamas (100% cotton) and to lie in a supine position on the bed with the electrodes placed under the bed sheet. In order to change BP, the subject was instructed to rest for 20 seconds, perform a Valsalva maneuver for 15 seconds, and then rest again for 90 seconds (total 125 seconds). Reference continuous BP (BPref) waves were simultaneously measured using a commercial non-invasive BP monitor (Finometer MIDI, Finapres Medical Systems). The subject wore a cuff sensor on the left middle finger. In order to keep the intrathoracic pressure constant during the Valsalva maneuver, the subject was asked to maintain the output of a digital manometer (HT-1500NM, HODAKA) at 40 mmHg. The measurement for a total of 125 seconds was used as a data set of one Valsalva test (VT), and the VT was repeated twice by each subject. During the first (1st VT) and second VTs (2nd VT), the subject was asked to maintain his leg position so that the measurement position would not shift.

**2.3 Signal processing**

Figure 2(a) shows the raw recordings of cECG, cBCG and BP_{ref} for the resting segment, 45 s after the Valsalva maneuver in the 2nd VT performed by subject #C, while (b) shows the enlarged waveforms of cECG, cBCG and BP_{ref}. The following signal processing was applied to an
80-s segment of simultaneously recorded cECG, cBCG and BPref signals, 45 to 125 seconds from the start time of measurement for each VT. Programs in MATLAB (Math Works) were used for the processing.

### 2.3.1 Preprocessing and extraction of parameters

The cECG was smoothed using moving-average filter with 20-ms (50 Hz) time window, then preprocessed by zero-phase band-pass filtering (finite impulse response filter, passband: 2.5–5.5 Hz) [20] and by differential processing. The preprocessed cBCG was used to detect the time of J-wave bottom (tBCG-J [ms]) within the range of 100–450 ms after tECG-R, and the bottom amplitude of J wave (VBCG-J) were determined as the amplitude at tBCG-J. Also, the time of K-wave peak (tBCG-K [ms]) and its amplitude (VBCG-K) were determined within the range of 300 ms behind tBCG-J. As shown in Fig. 3, VBCG was calculated as follows:

$$ V_{BCG} = V_{bcg-J} + V_{bcg-K} \quad (15) $$

Then, the zero-cross time (tBCG-ZC [ms]) was calculated between tBCG-J and tBCG-K as indicated in Fig. 3. Finally, pulse beat arrival time (TPBAT [ms]) was calculated as below:

$$ TPBAT = t_{bcg-ZC} - t_{bcg-R} \quad (16) $$

The BP ref signal was preprocessed by zero-phase low-pass filtering (finite impulse response filter, f_0 = 50 Hz). The values of both SBP (P_{SBP ref} [mmHg]) and diastolic BP (P_{DBP ref} [mmHg]) were obtained for each beat to calculate the reference MAP (P_{MAP ref} [mmHg]).

$$ P_{MAP ref} = P_{SBP ref} + \frac{P_{SBP ref} - P_{DBP ref}}{3} \quad (17) $$

### 2.3.2 Rejection of deviant cBCG waveform

Since BCG signal is susceptible to the subject’s body motion, deviant cBCG waveform and the corresponding parameters were rejected by the following process. First, an ensemble average of cBCG (Mbcg(\Delta t)) was calculated for each subject in synchronization with tBCG-J of every beat.

Next, the root mean square error (D_{RMSE, bcg} [mmHg]) was calculated in ±200 ms from tBCG-J and used for the outlier judgment. D_{RMSE, bcg} was calculated for each beat using the value of the preprocessed cBCG (v_{bcg(\Delta t)}) and...
\(M_{BCG}(l\Delta t)\) as follows:

\[
D_{RMSE,BCG} = \frac{\sum_{l=1}^{L}(V_{BCG}(l\Delta t) - M_{BCG}(l\Delta t))^2}{2L + 1}
\]

where \(\Delta t\) is sampling interval (1.0 ms), \(l\) is sample number, and \(L = 200\) in this study.

Finally, an outlier determination using a boxplot was used for the rejection processing. Using the first and third quartiles of \(D_{RMSE,BCG}\) (\(Q_1\) and \(Q_3\)), \(D_{RMSE,BCG}\) was labeled as an outlier as below [27]:

\[
D_{RMSE,BCG} < Q_1 - 1.5(Q_3 - Q_1)
\]

or

\[
D_{RMSE,BCG} > Q_1 + 1.5(Q_3 - Q_1)
\]

The labeled \(D_{RMSE,BCG}\) and corresponding parameters were rejected.

### 2.3.3 MAP estimation procedure

Flowcharts showing MAP estimation using (a) the conventional method and (b) the proposed method are shown in Fig. 4. In the proposed method [Fig. 4(b)], the estimation parameters \((V_{BCG}, N_{HR}, T_{PBAT}, P_{MAP,ref})\) were detected for each beat from the preprocessed cECG, cBCG and BPr. Next, for the estimation parameters, rejection of outlier was performed using \(D_{RMSE,BCG}\) (shown in 2.3.2). The estimation parameters were calculated using the moving average values for the 20-s window [28] from 45 to 125 s while shifting for each beat. After that, the estimation parameters were separated into a training segment (Tr, from 65 to 113 s) and a test segment (Te, from 113 to 125 s). To determine each parameter \((a\) to \(d)\) in Equation (12), we substituted \(P_{MAP,ref}\) on the left side and \(V_{BCG}, N_{HR}\) and \(T_{PBAT}\) on the right side for the Tr, and used the least squares method. The parameter values \((a\) to \(d)\) determined, and \(V_{BCG}, N_{HR}\) and \(T_{PBAT}\), were input into Equation (13), and \(P_{MAP,est}\) was calculated for the Tr and Te separately. In this fundamental study aiming at relative MAP estimation derived from a well-known equation about MAP, we used the same values of \(V_{BCG}, N_{HR}\) and \(T_{PBAT}\) for parameter determination and MAP estimation for the Tr and compared with the conventional method using only \(T_{PBAT}\). Moreover, we estimated the relative MAP for the Te for cross validation.

As shown in Fig. 4(a), the conventional method did not use \(V_{BCG}\) and \(N_{HR}\). Other than that, the procedure of the conventional method is the same as that of the proposed method.

### 2.4 Evaluation methods

To evaluate and compare the conventional and the proposed estimation methods, the following indices were calculated for each of the subjects: root mean square error (RMSE) between estimated and reference MAP, CC, and 95% confidence interval (CI). First, the RMSE \((E_{RMSE} [\text{mmHg}])\) was calculated as follows:

\[
E_{RMSE} = \sqrt{\frac{\sum_{l=1}^{N}(P_{MAP,est,l} - P_{MAP,ref,l})^2}{2N}}
\]

where \(N\) is the number of data.

Second, except for subject #G, 1st VT by the conventional method, CC \((r)\) was calculated using Spearman’s rank correlation coefficient. And, for subject #G, 1st VT by the conventional method, normality was detected for both \(P_{MAP,ref}\) and \(P_{MAP,est}\); therefore CC \((r)\) was calculated using Pearson’s product–moment correlation coefficient.

Finally, CI \((E_{CI} [\text{mmHg}])\) was calculated using the average of the reference and estimated MAP \((\mu [\text{mmHg}])\) and standard deviation of the reference and estimated MAP \((\sigma [\text{mmHg}])\) as follows:

\[
E_{CI} = (\mu + 1.96\sigma) - (\mu - 1.96\sigma)
\]

Significance tests on RMSE, CC, and CI for all subjects were performed using commercially available software (JMP®14, SAS Institute Inc., Cary, NC, USA).

### 3. Results

#### 3.1 Time series plots and RMSE

Figure 5 shows the temporal changes of \(P_{MAP,ref}\), estimated MAP by the conventional method \((MAP_{cnv})\), and estimated MAP by the proposed method \((MAP_{pro})\) in subject #B for (a) 1st VT and (b) 2nd VT. Here the subscript “cnv” or “pro” in each parameter indicates that the parameter was obtained by the “conventional” or “proposed” method. Additionally, the subscript “Tr” or “Te” in MAPest and MAPpro indicates that the estimation was performed for the “Training” or “Test” segment. As shown in Fig. 5, the values estimated by the proposed method were closer to MAPref than those estimated by the conventional method. Furthermore, in subject #B, we confirmed that the proposed method tracked the changes in MAP, not only during a monotonic decrease, but also in the case of a change from a decrease to an increase. The RMSE for the Tr \((RMSE_{Tr})\) by the proposed method for the 1st VT was 0.39 mmHg, which was 75.5% less than by the conventional method, and that for the 2nd VT was 0.35 mmHg, which was 56.3% less than by the conventional method. Figure 6(a) shows the comparison of RMSE\(_{Tr}\) between the conventional and proposed methods in individual subjects, while (b) shows the comparison of the group averaged RMSE\(_{Tr}\) between the conventional and proposed methods. From Fig. 6(a), the RMSE\(_{Tr}\) of individual subjects by the proposed method were less than 2.0 mmHg, and were lower than those by the conventional method. From Fig. 6(b), the group averaged RMSE\(_{Tr}\) by the proposed method was significantly lower than that by the conventional method \((p < 0.01, paired t-test)\).
3.2 Correlation

Figure 7 shows the correlation plots of MAP_ref versus MAP_est in subject #B for the (a) 1st VT and (b) 2nd VT. The regression line is drawn for the Tr. As shown in Fig. 7, MAP_est by the proposed method was closer to the regression line than that by the conventional method.
Furthermore, \( r \) for \( \text{Tr} \) by the conventional method were 0.76 and 0.53, whereas those by the proposed method were 0.96 and 0.88, which were higher. Figure 8(a) shows the comparison of \( \text{CC}_\text{Tr} \) for the \( \text{Tr} \) (\( \text{CC}_\text{Tr} \)) between the conventional and proposed methods in individual subjects, while (b) shows the comparison of the group averaged \( \text{CC}_\text{Tr} \) between the conventional and proposed methods. From Fig. 8(a), \( \text{CC}_\text{Tr} \) by the proposed method was higher than that by the conventional method in six of seven subjects. In subject #C, however, \( \text{CC}_\text{Tr} \) was lower by the proposed method than by the conventional method. From Fig. 8(b), the group-averaged \( \text{CC}_\text{Tr} \) by the pro-
posed method was significantly higher than that by the conventional method \( p < 0.01, \) Wilcoxon signed-rank test.

3.3 Bland-Altman plots and CI

**Figure 9** shows the Bland-Altman plots of MAP_{ref} and MAP_{est} in subject \#B for the (a) 1st VT and (b) 2nd VT. As shown in **Fig. 9**, in subject \#B, \( \mu \) and CI for the Tr (CI_{Tr}) of 1st and 2nd VT were smaller by the proposed
method than by the conventional method. Figure 10(a) shows the comparison of CI$_{Tr}$ between the conventional and proposed methods in individual subjects, while (b) shows comparison the group averaged CI$_{Tr}$ between the conventional and proposed methods. From Fig. 10(a), CI$_{Tr}$ by the proposed method were less than 7.5 mmHg in all subjects, which were lower than those by the conventional method. From Fig. 10(b), the group averaged CI$_{Tr}$ was significantly smaller by the proposed method than by the conventional method ($p < 0.01$, paired $t$-test).

**Fig. 9** Bland-Altman plots of MAP$_{ref}$ and MAP$_{est}$ in subject #B for the (a) 1st VT and (b) 2nd VT. Confidence interval for the Tr (CI$_{Tr}$) was calculated using values of MAP$_{est}$ and MAP$_{ref}$. Subscript "cnv" or "pro" in each parameter indicates that the parameter was obtained from the "conventional" or "proposed" method. Additionally, the subscript "Tr" or "Te" in MAP$_{est}$ and MAP$_{pro}$ indicates that the estimation was performed for "Training" or "Test" segments. The values of $\mu$, $\mu \pm 1.96\sigma$, and $N$ were obtained from the Tr for each method.

**Fig. 10** (a) Comparison of CI$_{Tr}$ calculated from MAP$_{est}$ and MAP$_{ref}$ between the conventional and proposed methods in individual subjects. (b) Comparison of group averaged CI$_{Tr}$ between the conventional and proposed methods. Double asterisks (**) indicate statistical significance at $p < 0.01$ based on paired $t$-test.
3.4 Cross validation

As shown in Fig. 5, for the Te, MAPpro was closer to MAPref than MAPcnv, but MAPpro partially showed an inverse correlation with MAPref. The RMSE of MAPpro for Te (RMSE_{Te}) for the 1st VT was 0.41 mmHg, which was 74.2% less than that of MAPcnv, and RMSE of MAPpro for the 2nd VT was 1.24 mmHg, which was 38.6% less than that of MAPcnv. Figure 11 shows the comparison of RMSE_{Te} between the conventional and proposed methods in individual subjects, while (b) shows the comparison of the group averaged RMSE_{Te} between the conventional and proposed methods. From Fig. 11(a), RMSE_{Te} by the proposed method was smaller than that by the conventional method in 10 of 14 VTs. However, for the 2nd VT in subjects #E and #F, it should be noted that RMSE_{Te} of the proposed method may be larger than that of the conventional method. From Fig. 11(b), the group averaged RMSE_{Te} was not remarkably different between the conventional and proposed methods. There was no significant difference in RMSE_{Te} between the conventional and proposed methods (Wilcoxon signed-rank test).

4. Discussion

For the Tr, the proposed method yielded significantly larger CC_{Tr} and significantly smaller RMSE_{Tr} and CI_{Tr} compared to the conventional method. These results for the Tr suggest that the proposed method produced results closer to the MAPref than did the conventional method. Furthermore, from Fig. 6(a) and Fig. 8(a), we confirmed that CC_{Tr} was not high even when RMSE_{Tr} was small, as shown by the 2nd VT in subject #B. For this reason, we considered that the 2nd VT of subject #B had a small MAPref fluctuation range for the Tr. The average MAPref fluctuation range for the Tr in all subjects was 12.5 ± 5.6 mmHg, while that for the 2nd VT in subject #B was the smallest at 3.1 mmHg, as shown in Fig. 5(b) and Fig. 7(b). Even when the fluctuation range was small as in subject #B, the proposed method had a higher CC_{Tr} and smaller RMSE_{Tr} than the conventional method. Additionally, since experimental data measured on a bed [20] was used for MAP estimation, there is a possibility that the proposed method for the Tr may be used to monitor the relative MAP of a supine and resting user in a cuffless, unrestrained, nonwearable and noncontact manner. In addition, RMSE_{Te} was smaller by the proposed method than by the conventional method in 10 of 14 data. Although Te had a short duration, it is possible that the proposed method may estimate MAPref with smaller RMSE_{Te} than the conventional method, even with unknown data. However, since the number of data for Te was small, we considered that CC and CI were not reliable values. Therefore, it is necessary to prepare experimental data that has longer duration than the data used in the present analysis. Assuming that cBCG and cECG have been stable for a long time, Fig. 2 suggests that cBCG is disturbed even in short time measurement. Thus, cBCG measurement may be disturbed depending on the state of capacitive coupling between the body and the sensor. In the future, in order to measure stable cBCG for a long time, it will be necessary to improve the signal-to-noise ratio of the sensor and stabilization of the capacitive coupling between the body and the sensor (such as using a foot pillow with dents), and then apply the proposed method to cBCG and cECG measured for a long time and evaluate also using CC and CI.
We used the systems for the cBCG and cECG measurement as shown in Fig. 1. For cECG measurement, the electrical activity of the heart is measured and only time information is used as shown in Fig. 3. Therefore, the measurement position has no significant effect on cECG. On the other hand, for cBCG measurement, the time for mechanical displacement to travel is used. Therefore, the measurement position affects cBCG. In our system, we adjust the measurement position for each subject, and since the position of the sensor is not fixed, the measurement position of cBCG is expected to change for each measurement even in the same subject. Hence, in order to improve the reproducibility of measurement, it is desirable to be able to measure cBCG without adjusting the position individually.

We used cBCG and cECG measured from seven healthy males in their early twenties for MAP estimation. In the future, we aim to estimate relative MAP in diverse age groups including the elderly and in females. Therefore, it is necessary to consider whether the measurement system shown in Fig. 1 can be applied to users other than males in their early twenties. In that case, it is better to increase the sensitivity of the system. For example, Starr et al. [29] reported a negative correlation between the amplitude of BCG and age. Additionally, Lee [30] reported that the amplitude of BCG at rest was lower in females than in males. Thus, it is desirable to increase the sensitivity of cBCG measurement system in order to measure cBCG with small amplitudes in subjects such as elderly females. As a method of increasing the sensitivity of the cBCG measurement system, it may be achieved by increasing \( \frac{df}{dc} \) (where \( f \) is the oscillation frequency of the astable multivibrator and \( C \) is the coupling capacitance between the body and the sensor). The relationship between the oscillation frequency and the coupling capacitance is \( f = \frac{1}{2\pi\sqrt{1/LC}} \). Therefore, if both sides are differentiated by \( C \), it becomes \( \frac{df}{dc} = -\frac{1}{2\pi\sqrt{1/LC}} \). For this reason, to increase \( \frac{df}{dc} \), it is preferable to reduce the coupling capacitance. The coupling capacitance can be expressed by \( C = \varepsilon S/d \) (where \( \varepsilon \) is the permittivity, \( S \) is the electrode area, \( d \) is the distance between the body and the sensor). If the electrode area of the sensor is reduced, the coupling capacitance also becomes smaller.

As a result, the sensitivity is expected to increase.

From Table 1, all subjects who participated in the experiment had similar physical constitution (height: 1.61–1.71 m, weight: 50.0–67.5 kg). However, for the purpose of realizing relative MAP estimation for users with different physical constitutions, it is necessary to validate whether the proposed method can be applied to subjects with different physical constitutions. The system requires adjustment of the position of the cBCG sensor at the heel when the height changes. Hence, it is desirable to improve the system so that it can be adjusted more easily and in a short time. As mentioned above, if the dent of a foot pillow naturally determines the position of the heel, it may be able to adjust the position of the sensor with ease. As a sensor applied on the foot for BP estimation, Martin et al. [18] used a PPG sensor array in combination with a weighing scale for BCG measurement. Since they succeeded in estimating SBP using pulse transit time from BCG and PPG, a combination of our system with the PPG sensor array may contribute to estimate not only MAP but also SBP simultaneously. Additionally, PPG sensor is clinically popular as an easy-to-wear sensor and is incorporated in the pulse oximeter. The pulse oximeter is used for screening sleep apnea syndrome (SAS), which is a type of sleep disordered breathing (SDB). As SDB is frequently associated with hypertension, ischemic heart disease and heart failure [31, 32], the combination of our system with a pulse oximeter would increase the clinical value of the current screening for SAS.

Using the proposed method, RMSE\(_{Te}\) was smaller than using the conventional method in 10 out of 14 data sets, but RMSE\(_{Te}\) for the 1st VT of subject \#D, the 2nd VT of subject \#E and the 2nd VT of subject \#F were more than 2-fold larger than the conventional method. However, to estimate relative MAP with high accuracy even for \( Te \), it is necessary to apply the proposed method to long measurement data and investigate the relationship between the time interval, amplitude and parameters (\( a \) to \( d \)) used for the estimation. For long measurement, it is necessary to improve the measurement system as discussed above. Table 2 shows the parameter values determined by Tr for the 1st and 2nd VTs of individual subjects. For the proposed method, the parameter \( c \) for the 1st VT of subject \#D was higher than that for other subjects, and parameter \( c \) for the 2nd VT of subject \#F was close to 0. Hence, overfitting with very large or small influence of \( T_{PBAT} \) is expected to increase RMSE\(_{Te}\). On the other hand, the cause of increased RMSE\(_{Te}\) in subject \#E could not be found from the signal trends and parameter values. Further analysis is required by applying the proposed method to long measurement data. The proposed method is presumed to cause overfitting because the parameters were not limited and the degree of freedom when determining the parameters was too high. As described above, overfitting may be less likely to occur if the parameters are determined after setting a limit such that each parameter is 0 or more.
Table 2  Parameter values determined for Tr and used in MAP estimation for Tr and Te (for each VT of each subject).

| subject | VT  | a   | b   | c   | d   | RMSE_{Te} |
|---------|-----|-----|-----|-----|-----|-----------|
|         |     | cvn pro | cvn pro | cvn pro | cvn pro | cvn pro |
| #A      | 1st | — | — | — | — | 1.28 | 2.76 | 1.5 | 20.2 | 13.7 | 2.41 | 1.56 |
|         | 2nd | — | — | — | — | 0.76 | 2.06 | 1.76 | 16.3 | 14.9 | 3.20 | 0.34 |
| #B      | 1st | — | — | — | — | 0.93 | 3.45 | — | 1.56 | 0.54 | 13.3 | 6.7 | 1.59 | 0.41 |
|         | 2nd | — | — | — | — | 0.71 | 0.34 | — | —0.81 | —0.76 | —0.3 | —1.3 | 2.02 | 1.24 |
| #C      | 1st | — | — | — | — | 0.18 | —0.07 | — | 2.08 | 4.34 | 15.9 | 24.2 | 1.52 | 0.30 |
|         | 2nd | — | — | — | — | 0.92 | —0.41 | — | 4.21 | 4.03 | 28.1 | 26.8 | 6.41 | 5.79 |
| #D      | 1st | — | — | — | — | 0.03 | 0.46 | — | 5.9 | 15.88 | 37.8 | 89.6 | 1.94 | 5.15 |
|         | 2nd | — | — | — | — | 1.83 | 0.4 | — | 9.2 | 5.56 | 56.5 | 39.7 | 3.49 | 1.94 |
| #E      | 1st | — | — | — | — | 1.08 | 1.57 | — | 3.61 | 3.65 | 24.9 | 25.6 | 2.17 | 2.39 |
|         | 2nd | — | — | — | — | 1.12 | —3.86 | — | 5.55 | 3.77 | 35.9 | 25.6 | 1.62 | 6.97 |
| #F      | 1st | — | — | — | — | 1.1 | —1.13 | — | 5.84 | 3.82 | 37.7 | 26.4 | 1.25 | 0.65 |
|         | 2nd | — | — | — | — | 1.76 | 0.26 | — | 1.08 | 0.04 | 10.6 | 8.2 | 0.53 | 5.32 |

5. Conclusion

In this study, we proposed and evaluated a method for cuffless continuous estimation of relative MAP using cBCG and cECG measured with an unrestrained and noncontact method. We adopted a well-known equation for calculating the MAP: MAP equals the product of CO and PVR. Then we derived an estimation formula for relative MAP using $V_{BCG}$ calculated from cBCG, $N_{HB}$ calculated from cECG, and $T_{PBAT}$ calculated from cBCG and cECG. We estimated MAP using the experimental data obtained from supine subjects who performed VT twice. We calculated and compared the RMSE, CC, and CI between the proposed method and the conventional method. Average RMSE_{Te} of MAP estimated by the proposed method for all subjects was 0.75 ± 0.33 mmHg, and was significantly lower than that by the conventional method. Average CC_{Tr} for all subjects by the proposed method was 0.93 ± 0.06, and was significantly higher than that by the conventional method. Average CI_{Tr} for all subjects by the proposed method was 2.96 ± 1.29 mmHg, and was significantly lower than that by the conventional method. Moreover, average RMSE_{Te} by the proposed method was 2.49 ± 2.22 mmHg. Although RMSE_{Te} was not significantly different between the conventional and proposed methods, smaller RMSE_{Te} by the proposed method was observed in 10 of 14 test segments. Therefore, these results indicate that continuous cuffless estimation of relative MAP using the proposed method can be used for short time measurement. As only males and a narrow age group were studied, further validation in diverse subjects is required in view of broad application of the proposed method.

Acknowledgment

This research was supported in part by SECOM Science and Technology Foundation and in part by Private University Research Branding Project supported by MEXT, Japan.

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