Direct Pulse Transit Time Measurement from an Electronic Weighing Scale

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Abstract

A novel method to measure Pulse Transit Time (PTT) from the force sensors of a modified electronic weighing scale is presented. The method relies on the relation between PTT and the interval between I and J waves of the ballistocardiogram (BCG). The IJ interval and the PTT derived from a carotid and a femoral pressure-pulse waveform were simultaneously obtained from three subjects during paced respiration, which induces periodic hemodynamic changes. The correlation between the IJ interval and this PTT is 0.69 and the difference between measurements is -5.2±16.4 ms, which are values similar to those of commercial devices based on more invasive methods that can be applied only to subjects in supine position. The simplicity of the method and the cost-effectiveness of the device to implement it offer new perspectives for the ambulatory assessment of arterial stiffness.

1. Introduction

Aortic stiffness and arterial stiffness in general are recognized independent predictors of cardiovascular disease (CVD) risk including myocardial infarction, heart failure, and total mortality, as well as stroke, dementia, and renal failure [1]. Pulse wave velocity (PWV), *i.e.* the propagation speed of the pressure pulse generated by cardiac ejection, is a common non-invasive method to evaluate arterial stiffness for CVD risk assessment [2] and has been included in recent guidelines for the management of arterial hypertension [3]. Since changes in arterial PWV can be detected in apparently asymptomatic patients before the development of CVD, its measurement in clinical and ambulatory settings is of great interest for screenings and periodic monitoring.

A simple and cost-effective method to measure PWV by sensors placed over large arteries more superficial and accessible than the aorta is from the pulse transit time (PTT), *i.e.* the time it takes the pressure pulse to travel from a proximal to a distal artery, divided by the estimated distance between the two measurement sites. The PTT measurement at the carotid-femoral segment is the simplest and most repeatable method to assess aortic stiffness and is the gold-standard due to its superior predictivity and reproducibility [4].

Pressure-pulse waveforms for PTT measurements can be measured either directly by high-fidelity applanation tonometers [5], or indirectly by mechanotransducers [6], photoplethysmographs (PPG) [7], and impedance plethysmographs (IPG) [8]. The proximal and distal waveforms can be obtained either simultaneously [6] or sequentially [5] by using the R wave of the ECG as a timing reference. These technologies are simple and cost-effective but inappropriate for fast screenings or periodic monitoring in non-clinical environments as the procedures require the exposure and preparation of the two measurement sites, which is tedious and time-consuming. Further, the sensors must be accurately placed over the desired arteries and this requires a trained operator.

An alternative method to obtain information about the propagation of the pressure pulse through the aorta is from the waves of the vertical ballistocardiogram (BCG). The BCG reflects the forces originated in the aorta as a result of cardiac ejection, and it can be obtained from the strain gages of common electronic weighing scales [9] thus obviating the placement of sensors on the body. The time interval between the R wave of the ECG and the J wave of the BCG, named RJ interval (see Figure 1) has been proposed as a surrogate of the cardiac ejection timing for the measurement of the pre-ejection period (PEP) [10] and related parameters such as cardiac contractility [11], which is inversely proportional to PEP in stable conditions of preload and afterload, and systolic blood pressure [12], whose fast variations are strongly correlated to PEP. However, the timing of the J wave with respect to other cardiovascular signals indicates that it is posterior to the onset of cardiac ejection and that it may therefore depend on the propagation of the pressure pulse through the arterial tree, as it is shown in the construct in Figure 1. On the other hand, earlier BCG waves that are closer to the beginning of cardiac ejection

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than the J wave could better coincide with relevant features of proximal pressure-pulse waveforms despite of its lower SNR. This has been suggested, for example, for the H wave [11], even though it precedes cardiac ejection and some authors have attributed it to ventricular movements due to its absence in recordings without ventricular contraction [13]. The I wave has also shown better results than the H and J waves when used as a surrogate of the pressure pulse at proximal sites [14] but a detailed analysis has not yet been reported.

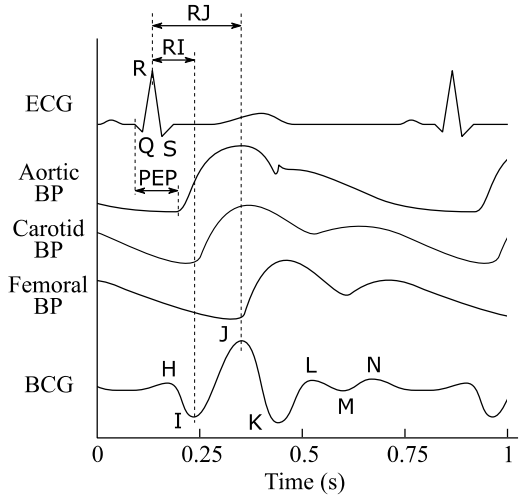

Figure 1. Diagram, based on bibliographic data, of the ECG, BCG, and the blood pressure traces at the root of the aorta, carotid artery, and the femoral artery with the respective main waves and time intervals of interest.

The different dynamics of the I and J waves with respect to the R peak of the ECG during paced respiration reported in [15] suggest that the interval between the I and J waves, named IJ interval, is sensitive to parameters other than the PEP, perhaps because of the dependence of the J wave on events posterior to cardiac ejection. Consequently, it is worth assessing to what extent the interval between I and J waves reflects changes in PTT. Since the measurement method only requires the subject to stand still on an electronic weighing scale, measuring the IJ interval in the BCG would better suit fast screenings or periodic monitoring than current methods based on sensors skillfully placed on specific parts of the body surface. This work describes a preliminary study of the relation between the IJ interval and the PTT.

2. Materials and methods

2.1. Signal acquisition systems

The four strain gages in an electronic weighing scale were connected into a Wheatstone bridge and interfaced to a differential amplifier with gain set to 5,000 by using a fully-differential, first-order, high-pass, passive filter [16], with corner frequency set to 0.5 Hz [17] to reject the body weight signal and low-frequency motion artifacts. The amplifier output was ac-coupled to an amplifier with gain 5 and corner frequency set to 0.5 Hz whose output was connected to a first-order, low-pass filter with corner frequency set to 25 Hz [17] to reduce high-frequency noise and power-line interference.

Since common devices used to measure carotid-femoral PTT [18] are designed to operate only on subjects in supine position, the reference PTT for our study was obtained from custom sensors able to measure in subjects that stand still on a weighing scale. The carotid pressure-pulse waveform was obtained from a piezoelectric force transducer (model 1010, UFI, Morro Bay, CA, USA), connected to a charge amplifier with 0.15 Hz to 22 Hz bandwidth according to the signal conditioning circuit proposed by the manufacturer. The femoral pressure-pulse waveform was obtained from a custom IPG system [19] that comprises a current injection circuit connected to two dry electrodes on the top surface of a weighing scale, intended to be contacted one by each foot, and a voltage detection system connected to two adhesive wet-gel (Ag/AgCl) electrodes. This approach minimizes artifacts when recording in standing subjects.

A standard ambulatory lead-I ECG was simultaneously recorded from adhesive wet-gel electrodes in order to provide a timing reference for the recorded waveforms.

2.2. Experimental setup

Signals were recorded from three healthy volunteers without any history of cardiovascular diseases (age: 28±4 years; gender: 3 males; weight: 72±11 kg; height: 181±10 cm) that gave their informed consent.

The sensors were interfaced to a computer via a data acquisition system (MicroDAQ-Lite, Eagle Technology, Cape Town, South Africa) configured to collect data from each channel at 1 kHz. The carotid sensor was held over the left sternomastoid muscle at the level of the larynx by an elastic band. The adhesive electrodes of the femoral IPG sensor were placed in the area around the mid-inguinal point, separated about 10 cm, and the two ECG electrodes were fixed each on the back of one hand. The subjects were asked to stand still in the modified weighing scale with their bare feet in contact with each of the two IPG electrodes.

The subjects first stood still for 20 s to let their hemodynamic parameters to stabilize. After that, signals were recorded for 100 s while the subjects were performing a paced respiration maneuver at 0.1 Hz in order to modulate the PTT without causing excessive motion artifacts.

2.3. Signal processing and data analysis

RJ intervals were obtained from the BCG by using the ECG as a timing reference. The beginning of individual heartbeats in the BCG was identified from the R waves of the ECG detected by a Pan-Tompkins algorithm [20]. To reduce the influence of unavoidable motion artifacts in BCG recordings from standing subjects, I and J waves were measured in a 3-heartbeat ensemble average that experimentally showed the best trade-off between noise reduction and waveform smoothening due to misalignment. The J wave was detected in each ensemble average as the absolute maximum in the interval from 100 ms to 250 ms after the R wave and the I wave was detected as the first local minimum before the J wave.

The reference PTT was obtained from the foot-to-foot interval in the IPG identified by the intersecting tangents in the carotid and femoral waveforms. The relationship between IJ intervals and the reference PTT was studied by the Bland-Altman analysis.

3. Results and discussion

Figure 2 shows a sample of waveforms collected from each subject, wherein the feet of the carotid and femoral pulses match the peaks of I and J waves respectively. In the aggregated analysis of the 407 heartbeats, I waves arrive 5.9±12.5 ms after the foot of the carotid pulse whereas J waves occur 0.7±21.0 ms after the foot of the femoral pulse, which is within the range of other time intervals derived from BCG signals [11][12].

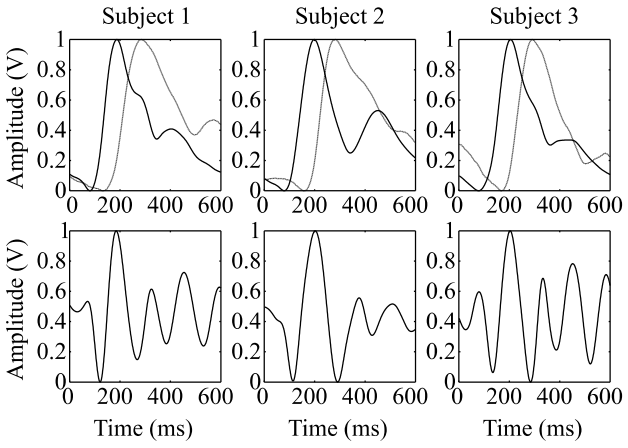

Figure 2. Sample of the carotid pulse (up, solid), femoral pressure-pulse (up, dotted), and BCG (down), collected from the three subjects.

Figure 3 shows a sample of the IJ interval and their corresponding PTT traces recorded during the experiments. The visual correlation between them is apparent.

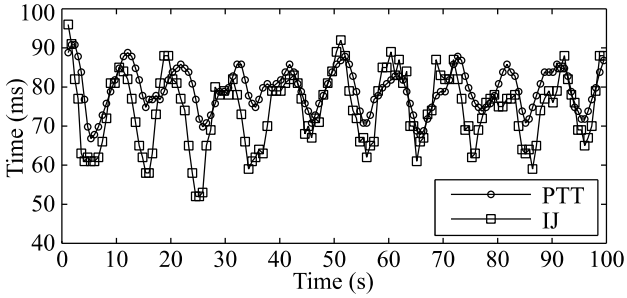

Figure 3. Sample of IJ interval and PTT traces recorded during paced respiration.

The modulation of the PTT caused by paced respiration is mostly in phase with the changes observed in the IJ interval although the variability in this last interval is higher, as expected from BGC recordings obtained from subjects that stand on a weighing scale without using any sensor in direct contact with their skin, which makes the signal more prone to motion artifacts.

The correlation coefficients between the IJ interval and PTT for each subject were 0.65, 0.62, and 0.65 ($p << 0.05$). The Bland-Altman analysis of the aggregated 407 pairs of values obtained from the experiments is shown in Figure 4.

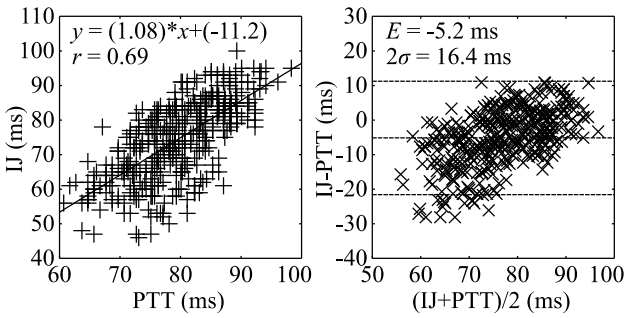

Figure 4. Bland-Altman analysis of the aggregated pairs of values obtained during paced respiration.

The Bland-Altman analysis reveals that the correlation is slightly stronger ($p << 0.05$) for the aggregated results than for the individual ones. The IJ interval is about 5.2 ms shorter than the PTT, with a standard deviation of 16.4 ms (for 2σ) and a slight bias that causes an underestimation of the PTT at lower values.

The synchronism between IJ interval and PTT traces during paced respiration and the correlation between them suggests that the IJ interval strongly depends on the PTT and their variations are concomitant. The results for the IJ interval are similar to those from widely-accepted devices that also rely on carotid-femoral PTT [18], such as Complior ($\rho = 0.78$, agreement 19.5±20.2 ms). While the agreement is in the same range, the better correlation obtained by these devices can be attributed to the deleterious effect of motion artifacts inherently present in

BCG recordings that hinder the measurement of the IJ interval and also to the wider degree of variations in PTT (from 30 m/s to 100 m/s), which is comparatively narrower in our experiments. Further, systems for aortic PWV measurement that do not rely on carotid-femoral PTT, such as the PulseTrace PCA system, are valued in clinical practice in spite of their lower correlation values (ρ = 0.55) with PWV measured from aortic PTT [18]. On the other hand, the IJ method is far less invasive and does not require any skilled operator, which can increase significantly the range of scenarios to monitor arterial elasticity compared to the current limited use and availability of this kind of measurements.

4. Conclusions

The time interval between I and J waves of the BCG obtained from the force sensors in a common electronic weighing scale is correlated to the carotid-femoral PTT. In measurements in three subjects that performed a paced respiration maneuver, the correlation was ρ = 0.69, with a mean difference of -5.2 ms and a standard deviation of 16.4 ms. These values are close to those from some commercial methods currently used to estimate PWV.

Even though the proposed method has been assessed only in three subjects, the results are very encouraging because it allows us to estimate PTT in clinical and non-clinical scenarios. The method only requires the subject to stand on an electronic weighing scale, or another platform if the weighing function is not desired, and does not need the user to expose any body part. The widespread use of electronic weighing scales and the need of alternative methods for screening and periodic monitoring suitable for ambulatory scenarios support further investigation including its evaluation for a wider and heterogeneous cohort.

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