Diastolic Augmentation Index Improves Augmentation Index in Assessing Arterial Stiffness

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Abstract

Arterial stiffness is an important risk factor for cardiovascular (CV) events and is increasingly used in clinical practice. Radial augmentation index (AI) is used in assessing arterial stiffness, but is not only dependent on pulse wave velocity (PWV), but also on several other factors like the reflect distance of pulse wave and height. This paper improved radial AI in assessing arterial stiffness by d-value, the subtraction between radial AI and diastolic AI (dAI). Nineteen subjects aged 18 to 80 years (mean±SD, 46±27 years) were enrolled in this study. Carotid-femoral PWV (cf-PWV), radial AI and dAI of each subject were measured. The d-value (r=0.69, P<0.005) shows better linearity with cf-PWV than AI (r=0.62, P<0.01) and dAI (not significant) do. In conclusion, dAI improves AI in assessing arterial stiffness.

1. Introduction

Arterial stiffness is an important risk factor for cardiovascular (CV) events and is increasingly used clinically [1-4]. Many indicators have been proposed for the assessment of arterial stiffness. These indicators are basically classified into three types: systemic, regional and local arterial stiffness. In addition, wave reflection is also of great interest to estimate arterial stiffness.

Systemic arterial stiffness is available through use of a modified Windkessel model [5, 6], by use of the 'area method' [7, 8] or is simplified as the ratio between stroke volume (SV) and pulse pressure. However, these methods all involve an electrical model which is considered an inappropriate approximation of the circulation in the assessment of arterial stiffness [9]. In addition, the calculation of SV based on a single pulse wave (especially one at a peripheral site) is not well validated.

Local arterial stiffness is directly available (calculated as the change in volume divided by the change in pressure) at a specific site of the artery. The local arterial stiffness of superficial arteries is usually measured using echo-tracking devices. The carotid stiffness wins particular interest as is close to the aorta and is a site where atherosclerosis often occurs. However, its application in clinical is limited by the need of technical expertise. In addition, the carotid stiffness cannot be used as a surrogate of the aortic stiffness in patients with high blood pressure and/or diabetes [10]. The local arterial stiffness of deep arteries such as the ascending aorta is available through use of cine magnetic resonance imaging. However, the precision in determining the change of arterial diameter should be improved [11].

Regional arterial stiffness is usually assessed by the pulse wave velocity (PWV) of a specific segment of artery. Carotid-femoral PWV (cf-PWV) is considered as the 'gold standard' determinant of arterial stiffness [1, 11]. However, limitations still exist. First, it is not convenient to record the carotid and femoral pulse waves simultaneously and patients should always keep in supine position. Second, the distance from the carotid to the femoral artery is difficult to measure accurately especially in patients with abdominal obesity [12]. In addition, the femoral pulse wave is not accurately available in patients with obesity, diabetes, metabolic syndrome, and/or peripheral artery disease [11].

Wave reflection is always used to estimate arterial stiffness. A pulse wave is composed of a forward and its corresponding reflected wave. The forward wave is generated by the pump of the heart, travels along the artery and is reflected at branch points or sites of impedance mismatch. The reflected wave then travels back and meets the forward wave. The reflected wave is added to the forward wave and cause augmentation in pulse pressure. This augmentation varies with PWV, which determines the time when the reflected wave meets the forward wave. When the artery is less stiff, the pulse wave velocity is low, the reflected wave meets the forward wave during diastole, whereas when the artery is stiffer, the reflected wave meets the forward wave during systole, highly augmented the pulse pressure. This

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augmentation caused by wave reflection is quantified by augmentation index (AI). Aortic AI has been shown to be an independent predictor of all-cause and CV mortality in end-stage renal failure patients [13]. Aortic AI corrected for heart rate (HR) of 75 bpm (AIx@75) has been proved to be independently associated with severe short- and long-term cardiovascular events in patients undergoing PCI [14]. However, aortic AI can hardly be obtained non-invasively. Radial augmentation index is conveniently available and is used to assess arterial stiffness in a broadly accepted device: HEM9000AI (Omron Healthcare, Japan). Radial augmentation index is shown to be predictive of CV events. Radial augmentation index is age-dependent and could be a useful index of vascular aging [15]. It is also reported that radial augmentation index is a predictor of premature coronary artery disease in younger males [16]. However, radial AI is not only dependent on PWV, but also on several other factors like the reflect distance of the pulse wave and height. In addition, it is shown that radial AI failed to measure vascular stiffness in the elderly over the age of 55 [17].

This paper improved radial AI in assessing arterial stiffness by the subtraction of radial AI and dAI. The subsequent contents are organized as follows: the second section describes the methodologies used in this study; the third section illustrates the results; the discussion and conclusion are demonstrated in the fourth section.

2. Methodology

2.1. Study protocol

Nineteem subjects aged 18 to 80 years (mean±SD, 46±27 years) were enrolled in this study. Detailed information and physiologic conditions of the subjects are shown in Table 1. Subjects have a 10 min rest before the acquisition. During the acquisition procedure, subjects were kept in supine position all the time.

Carotid and femoral pulse waves were measured simultaneously using pressure pulse sensors. The data were recorded at a sampling frequency of 1000Hz. The PWV distance was measured using a tape from the carotid to the femoral artery. The radial pulse wave was recorded using the SphygmoCor device (AtCor, Australia) with a sampling frequency of 128Hz. AI and dAI were calculated from the radial pulse wave. To improve the accuracy, measurements of cf-PWV, AI and dAI were all repeated three times in each subject. Then the mean of the three measurements were taken to determine cf-PWV, AI and dAI of each subject.

2.2. Data processing

The carotid and femoral pulse waves should be pre-processed to eliminate baseline drift and noise, which highly influences the accuracy of subsequent calculation. Baseline drift is mainly brought in by body motion artifact and respiration. Wavelet decomposition was employed in drift removal. Wavelet decomposition at level 10 was applied to the data and the approximation coefficients were eliminated. Similarly, the noise was removed by applying wavelet decomposition at level 4 to the data and eliminating the detail coefficients.

Table 1. Information of the subjects.

No.	Sex (year)	Age	Height (cm)	Weight (kg)	BP (mmHg)	HR (bpm)
1	F	23	161	48	102/68	66
2	F	26	157	44	90/62	64
3	F	19	172	61	118/70	66
4	F	18	167	62	125/75	74
5	M	19	175	69	116/70	60
6	M	19	177	63	115/75	66
7	M	18	177	67	120/80	74
8	M	19	181	64	135/75	73
9	M	79	163	65	140/78	80
10	F	60	160	60	120/64	59
11	F	65	160	47	112/72	80
12	M	80	175	75	138/82	66
13	M	64	174	64	126/78	86
14	M	71	170	67	146/70	53
15	M	76	165	53	106/54	61
16	F	80	160	63	134/60	54

2.3. Calculation of PWV

Carotid-femoral PWV was calculated by dividing travelled distance by pulse transit time (PTT). The distance was calculated as 0.8 times the direct distance from the right common carotid artery to the common femoral artery.

The PTT were calculated as the difference between the feet of the carotid and femoral pulse waves in time. The feet of the carotid and femoral pulse waves were both extracted using the intersecting tangents technique [18-20], which determines the foot by the intersection of the horizontal line through the minimum and the tangent line through the maximum first derivative with respect to time.

PTT was obtained from each cardiac cycle in a series of data. In order to improve the accuracy and robustness of PTT calculation, the one exceeded the 90% of the SD distribution curve of the PTTs was discarded. The remaining PTTs were then averaged.

2.4. Calculation of AI and dAI

As shown in Figure 1, AI is defined as the ratio between the second (P2) and first (P1) peak of radial pulse wave, and dAI is defined as diastolic peak (Pd) divided by P1. The difference between AI and dAI was named d-value, which means: d-value=AI-dAI.

AI was calculated automatically by the SphygmoCor device. dAI was calculated in each cardiac cycle. In each series of dAI derived from all cardiac cycles, the one exceeded the 90% of the SD distribution curve of dAI was discarded. The dAIs were then averaged to reduce the influence of respiration on the calculation of dAIs.

For each individual, the average radial pulse wave was derived using ensemble average method. Then, AI and dAI were both calculated from the averaged pulse wave.

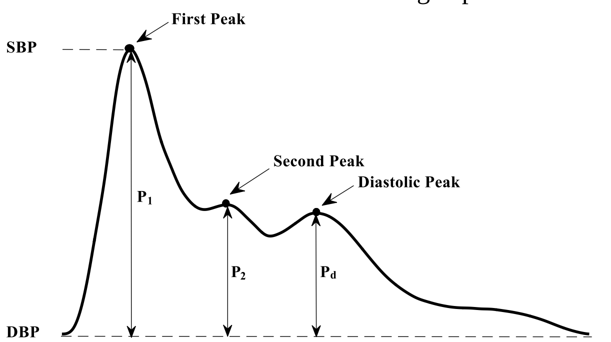

Figure 1. High-fidelity recording of the radial pulse wave. Amplitude of the peak and foot are the systolic (SBP) and diastolic (DBP) blood pressures, respectively. P_1 indicates the difference between the first peak and the foot in amplitude; P_2 is the amplitude of the second peak minus DBP; P_d is the amplitude of the diastolic peak minus DBP.

2.5. Statistical analysis

To study the improvement of dAI on AI in assessing arterial stiffness, the linearity between d-value and cf-PWV was compared with the ones between AI and cf-PWV and between dAI and cf-PWV, respectively. The linearity was evaluated by correlation coefficients.

3. Results

3.1. Reliability test

To test the reliability of all the measurements, correlation coefficient between the first and second measurements of each parameter (cf-PWV, AI or dAI) in all subjects was calculated. The correlation coefficient between the first and second measurements of cf-PWV was r=0.97, P<0.001. The correlation coefficient between the first and second measurements of AI were r=0.97, P<0.001 and r=0.87, P<0.001, respectively.

3.2. Linearity between d-value and PWV

Figure 2 illustrates the linearity between cf-PWV and AI, dAI and d-value, respectively. AI is significantly correlated with cf-PWV (r=0.62, P<0.005). dAI and cf-PWV shows no significant linearity. Compared with AI, d-value significantly correlates with cf-PWV with a greater correlation coefficient (r=0.68, P<0.001).

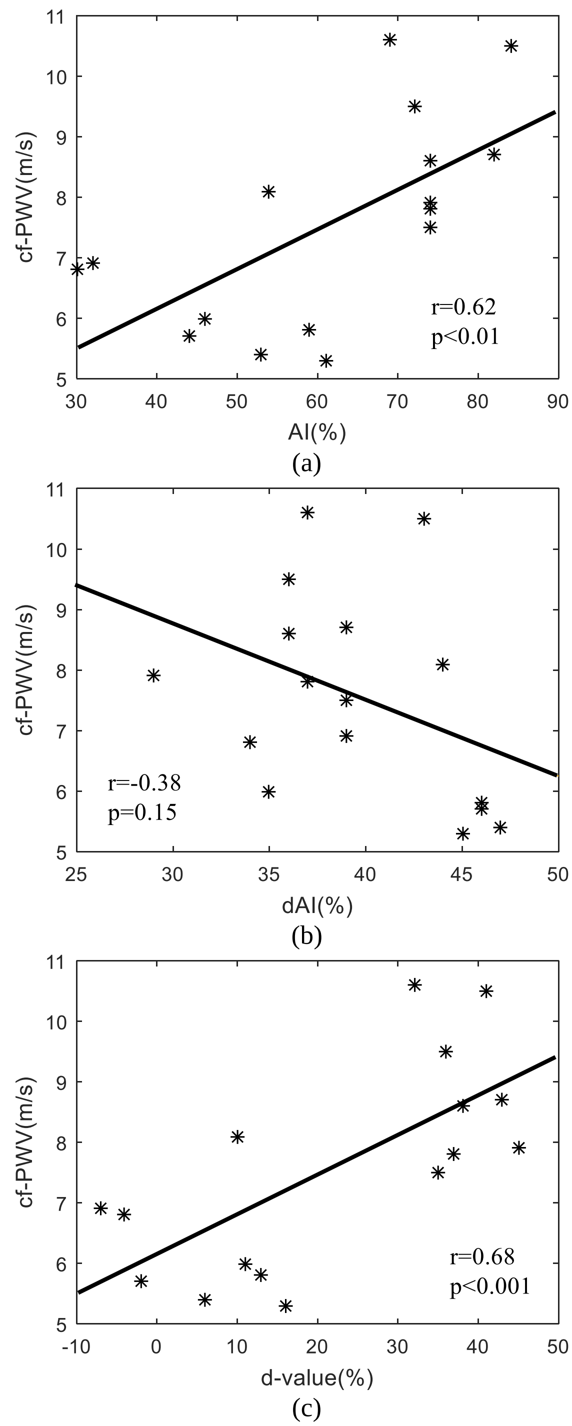

Figure 2. Performance comparison between AI, dAI and d-value in assessing arterial stiffness: (a) linearity between AI and cf-PWV; (b) linearity between dAI and cf-PWV; (c) linearity between d-value and cf-PWV.

4. Discussion and conclusion

This paper compared the performance of AI and d-value in assessing arterial stiffness by studying their linearity with cf-PWV. AI shows positive correlation (r=0.62, P<0.005) with cf-PWV whereas dAI shows no significant correlation with cf-PWV. d-value shows better linearity (r=0.68, P<0.001) with cf-PWV than AI does. In conclusion, dAI indeed improved AI in assessing arterial stiffness. Further study is needed on the relationship between d-value and CV events.

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