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Estimation of Local Aortic Elastic Properties With MRI

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Aortic compliance and pulse wave velocity (PWV) are important determiners of heart load, and are clinically useful indices of cardiovascular risk. Most direct methods to derive them require invasive pressure measurement. In this work a noninvasive technique to evaluate aortic compliance and PWV using MRI is proposed. MRI magnitude and phase images to measure area and flow in the ascending aorta were acquired in a group of 13 young healthy subjects. Assuming that the early systolic part of the wave was unidirectional and reflectionless, PWV was determined as the ratio between flow and area variations at early systole. Our results were compared to pulse wave velocities derived from a direct transit time, and to one using ascending aortic area and peripheral brachial pulse pressure. The new method proved to be accurate and in good agreement with the transit time method, as well as with previously published results. Magn Reson Med 47:649–654, 2002.

Key words: MRI; ascending aorta; pulse wave velocity; compliance; arterial wall

Reduced arterial compliance has been shown to correlate with age and pathologic states such as arteriosclerosis, coronary heart disease, and hypertension (1–6). The aorta is a major determiner of total systemic compliance, contributing to 60–70% of its total value (7). Elasticity of the aortic wall is mostly responsible for the buffering Windkessel effect, which limits excessive pressure pulsation in systemic arteries. Loss of aortic compliance has important physiopathological relevance. First, it leads to increased systolic and decreased diastolic pressure due to loss of buffering, thereby increasing the pulse pressure. Second, the higher pulse wave velocity (PWV) due to a stiffer wall leads to early arrival of reflection pressure waves in the aorta, which contributes to an increase in systolic and pulse pressure. Higher pulse pressure may be responsible for long-term medial damage, probably through increased cyclic stress and fatigue. It also leads to cardiac pressure overload and hypertrophy, and is known to be a major risk factor in coronary heart disease (8). Moreover, the lower diastolic pressure leads to a reduction of the coronary perfusion. For the above-mentioned reasons, the precise measurement of the local aortic compliance may increase understanding of arterial physiopathology, and become a clinical marker for cardiovascular risk and a useful tool for treatment monitoring.

The direct determination of local aortic compliance requires the measurement of aortic diameter and pulse pressure. Some previous studies obtained aortic pressure with invasive intravascular local measurements (1,3,9,10), while others used brachial pressure measured by a sphygmomanometer as a surrogate for aortic pressure (1,4–6).

We propose a new, noninvasive method to evaluate compliance and the related PWV of the ascending aorta. It derives compliance from local early systolic flow and cross-sectional area and does not ask for central pressure values. Flow and area are obtained from MRI measurements. To assess the validity of this method, its results are compared to those of another MRI method (5), which measures wave velocity across the aortic arch, as well as to a third method based on aortic cross-sectional area variations and brachial pulse pressure.

THEORY

PWV and Compliance Estimation From Early Systolic Flow and Area (QA Method)

The method for calculating the PWV and compliance (C) is based on the following considerations: during early systole, aortic pressure and flow waves do not contain reflection waves, since reflected waves take a finite time to reach the aorta from the periphery. We can therefore assume early systole to be unidirectional and reflectionless (11). For unidirectional waves, the ratio between pressure variation (ΔP) and flow variation (ΔQ) is then equal to the characteristic impedance Z_C:

$$Z_C = \frac{\Delta P}{\Delta Q}. \quad [1]$$

By definition, the local area compliance C_A is given by

$$C_A = \frac{\Delta A}{\Delta P} \quad [2]$$

where ΔA stands for the variation of cross-sectional area. C_A is related to Z_C through the following formula:

$$Z_C = \sqrt{\frac{\rho}{A}\frac{1}{C_A}} \quad [3]$$

where ρ stands for the blood density and A for the cross-sectional area at the end diastole. Eliminating ΔP from the above expressions yields the following expression for C_A:

$$C_A = \left(\frac{\Delta A}{\Delta Q}\right)^2 \frac{A}{\rho}. \quad [4]$$

C_A is related directly to PWV (PWV_{QA}) by

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$$PWV_{QA} = \sqrt{\frac{A}{\rho}\frac{1}{C_A}}. \quad [5]$$

Therefore, by substituting Eq. [4] for C_A we obtain:

$$PWV_{QA} = \frac{\Delta Q}{\Delta A}. \quad [6]$$

PWV Using Transit Time of the Foot of the Flow Wave (TT Method)

In a second method (5), PWV is calculated using the transit time (Δt) of the foot of the flow wave across the aortic arch and the distance (Δx) between the locations of both measurements. The PWV_{TT} is then given by:

$$PWV_{TT} = \frac{\Delta x}{\Delta t}. \quad [7]$$

The above expression applies to flow wave velocity, which equals PWV if the fluid is incompressible and the wall nonviscous.

PWV and Compliance Using Aortic Area and Brachial Pulse Pressure (BR Method)

Ideally, the easiest and most direct way to determine compliance would be to determine the ratio between the area variation and central pulse pressure, but, as described above, estimation of the central pulse pressure is technically very demanding or invasive. Using brachial pulse pressure (PP_{BR}) as a surrogate for the central pressure, we calculate:

$$C_{BR} = \frac{\Delta A}{PP_{BR}} \quad \text{and} \quad PWV_{BR} = \sqrt{\frac{A}{\rho}\frac{1}{C_{BR}}}. \quad [8]$$

METHODS

Subjects

The study involved 10 young to middle-aged healthy men and three young healthy women (range 26–48, mean age 34.3) with no symptoms of cardiovascular disease or hypertension (mean brachial systolic pressure 113.3 mmHg, range 105–123 mmHg; mean brachial diastolic pressure 65 mmHg, range 53–79 mmHg).

MRI Measurements

Measurements were obtained in a 1.5 T clinical imager using the body coil with ECG gating. Flow and area in the ascending and descending aortas were measured using a time frame of 8 ms by one double-oblique slice perpendicular to the aorta at about 5 cm above the aortic valve. For flow, a set of 24 fast low-angle shot (FLASH) 2D sequences (TR = 24 ms, TE = 5 ms, flip angle = 30°, matrix size = 256 × 256, pixel size = 1 × 1 × 6 mm, acquisition time < 4 min) with flow-sensitive phase images encoded up to 250 cm/s were repeated three times over 256 heartbeats with a trigger delay of 0, 8, and 16 ms from the QRS sampling 576 ms of the cardiac cycle. For the area measurements, a set of eight segmented FLASH sequences (TR = 64 ms, TE = 6.1 ms, flip angle = 30°, matrix size = 130 × 256 , three averages, pixel size = 2 × 1 × 6 mm, five *k*-space lines acquired at each excitation, acquisition time < 1 min 30 s) was repeated eight times over 78 heartbeats at 0, 8, 16,. . .,56 ms trigger delay from the QRS. This sequence was chosen to give the best delineation of the aortic wall. The whole process produced two sets of images (intensity and phase), as shown in Fig. 1.

Brachial systolic, diastolic, and mean blood pressures were measured by means of a sphygmomanometer every 2 min in the course of the MR acquisition.

Data Processing

For each picture, the lumen of the ascending and the descending aorta in the intensity set of images was outlined on the computer screen using the NIH Image software (Fig. 1). By transferring this outline to the phase images, the flow was obtained by adding the volume flow of each pixel inside the aorta. The value of the aortic cross-section and of the blood flow in both the ascending and the descending aorta was available every 8 ms during the heart cycle. To reduce variation, the area was measured twice and the mean area was used. The relative error of the area measurement was evaluated by outlining both a good-quality and a poor-quality image 10 times, rating subjectively, and calculating the standard deviation (SD) divided by the mean value.

Using the TT method, processing was similar to the method described initially by Mohiaddin et al. (5). The distance Δx between the two aortic cross-sections represented on the transverse image was measured manually across the aortic luminal midline on a double-oblique image in the plane of the aortic arch. The foot of the ascending and descending aorta flow waves was estimated as the point of interception of the linear extrapolation of the steep early systolic slope and the late diastolic flow baseline (Fig. 2).

The difference between maximal systolic and end diastolic areas was used to calculate PWV_{BR}. Pulse pressure was given by systolic minus diastolic brachial pressure.

Two-tailed paired *t*-tests were performed to test the difference between the three methods. A Bland-Altmann plot (12) was used to check the agreement between PWV_{QA} and PWV_{TT} and to exclude a specific trend in the distribution.

RESULTS

In our group of 13 healthy subjects, the mean area was 7.2 ± 1.9 cm^2 and the mean variation of the area was 18% (range 8–35%). The relative error on each area measurement ranged between 4% and 6%, depending on the image quality and on subject collaboration. Maximal flow amounted to 410 ± 60 ml/s. Flow and area measurements during systole and early diastole in a healthy volunteer are shown in Fig. 3a.

Figure 3b shows area plotted vs. flow for a healthy volunteer. At early systole, the data lie upon a straight line (fitted curve for the slope $\Delta Q/\Delta A = PWV_{QA}$), which supports the hypothesis of absence of reflection in this part of the wave that was assessed in the theory section.

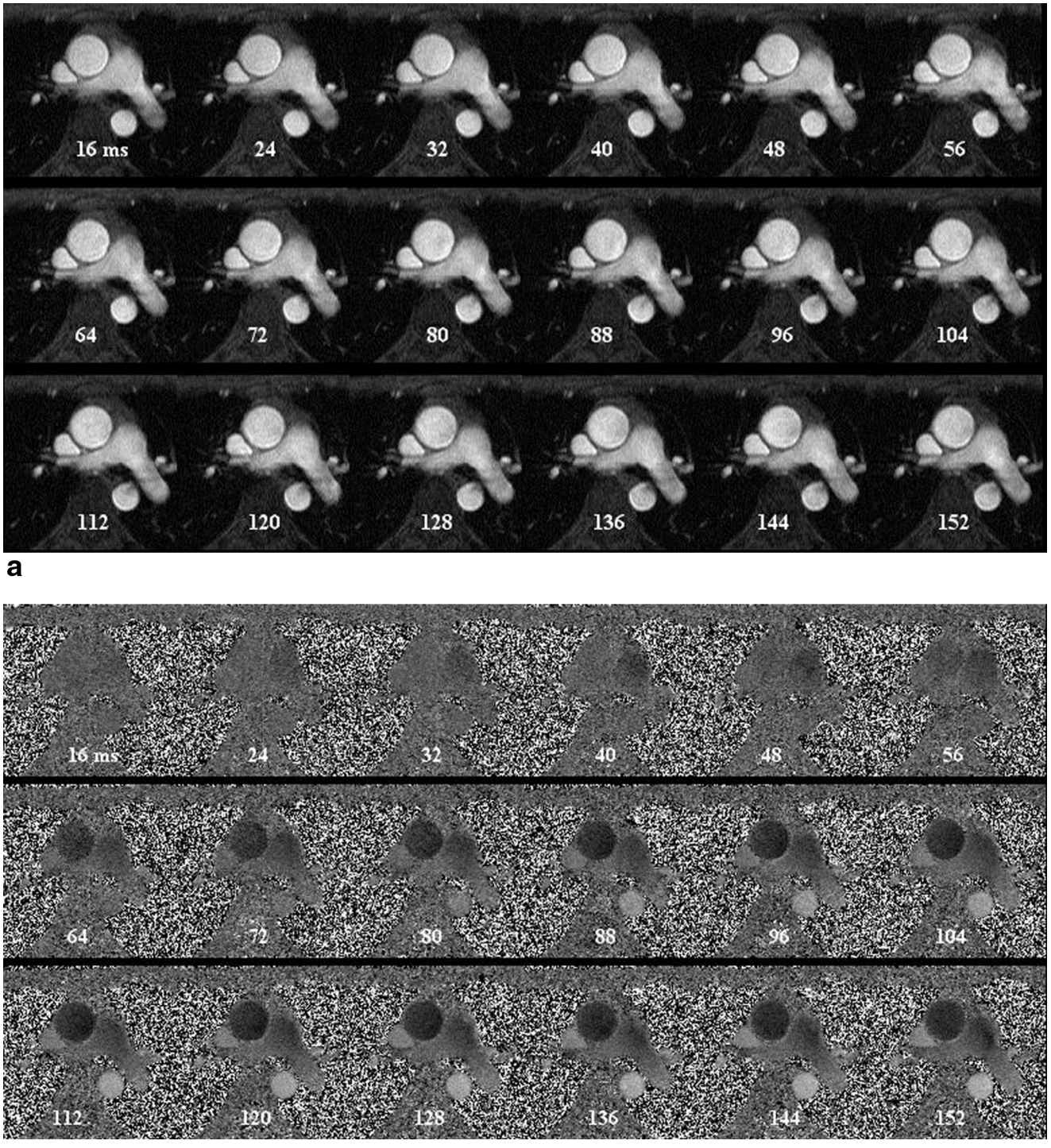

FIG. 1. MR double-oblique images showing the magnitude series used to measure (**a**) the area and (**b**) the phase series used to measure the flow as acquired in the ascending and descending aorta during early systole. Time delay of the acquisition from the R-wave of the ECG is mentioned at the bottom of each image.

The mean values (SDs) of PWV according to the three methods were PWV_{QA} = 4.9 (1.1) m/s, PWV_{TT} = 4.4 (0.9) m/s and PWV_{BR} = 6.7 (1.4) m/s. The paired *t*-test showed that the PWV_{QA} and PWV_{TT} values were not statistically different ($P > 0.05$) but that PWV_{QA} was significantly lower than PWV_{BR} ($P < 0.0001$). Figure 4 shows a comparison of the PWVs for all subjects. PWV_{QA} and PWV_{TT} are close together and lower than PWV_{BR} for every subject.

A Bland-Altman (12) plot for PWV_{QA} and PWV_{TT} is given in Fig. 5, with mean value and SD of the difference between PWV_{QA} and PWV_{TT} of 0.5 $\pm$ 1.0 m/s. As mentioned for Fig. 4, the positive difference is inconsistent and

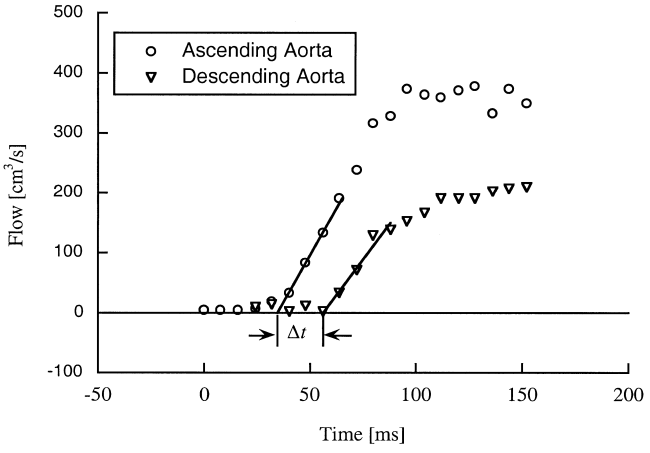

FIG. 2. Flow in the ascending and descending aorta during systole. Measurement of the foot-to-foot time delay yields PWV_{TT}.

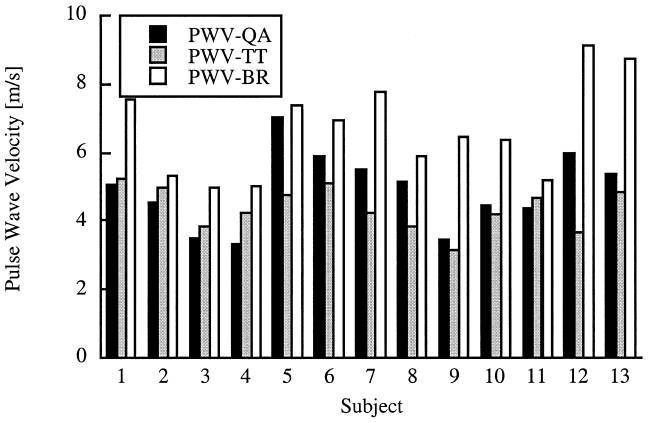

FIG. 4. PWV for all subjects obtained from the flow area (PWV_{QA}), transit time (PWV_{TT}), and brachial pressure methods (PWV_{BR}).

the differences are clustered around the zero value. No dependence on PWV is observed in the distribution.

DISCUSSION

Compliance of the aorta is an important determiner of heart load and a clinically useful index to assess cardiovascular risk, because the aorta is the major contributor to total systemic compliance. To calculate aortic compliance, local area and pressure measurements are needed, the latter being only obtainable by invasive means (1,9,10). A new method is proposed here, to derive aortic compliance and PWV based on local flow and cross-sectional area measurements using MRI.

PWV values of the ascending aorta obtained in our study are in good agreement with other MRI studies in healthy volunteers. Mohiaddin et al. (5) found PWV values between 4.3 ± 0.7 m/s and 7.2 ± 0.2 m/s, depending on the age of the subject group, while Groenink et al. (2) found PWV values of 3.9 ± 0.4 m/s. Using transthoracic echocardiography, Jeremy et al. (13) determined a value of 3.9 ± 0.6 m/s. Further investigations of descending thoracic or abdominal aorta using transthoracic ultrasound (3,14,15), transoesophageal ultrasound (16), or intravascular ultrasound (10) reported higher PWV or lower compliance, which is consistent with cross-section reduction and stiffness increase when moving distally. The comparison with previous studies shows that the PWV of the ascending

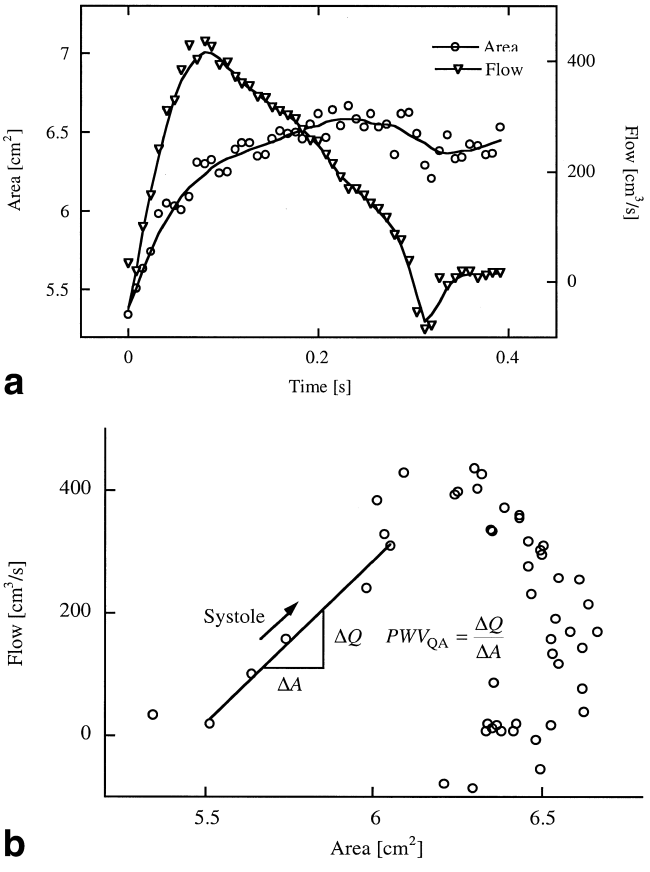

FIG. 3. **a:** Area and flow in the ascending aorta during systole and early diastole. **b:** Flow plotted vs. area. The slope of a straight line fitted to the early systole gives $\Delta Q/\Delta A = PWV_{QA}$.

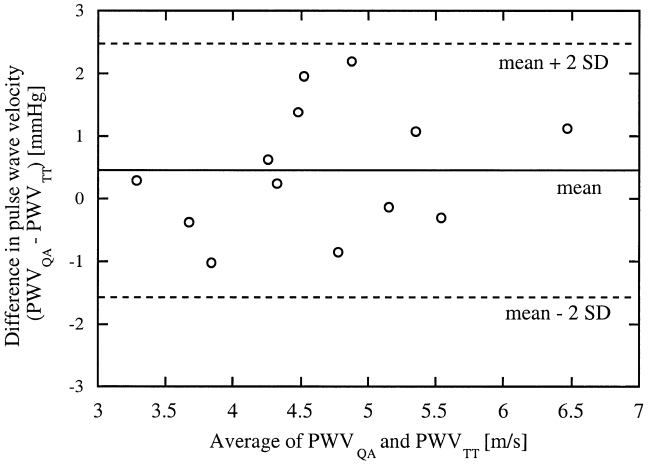

FIG. 5. Bland-Altman plots for the comparison of PWV_{QA} and PWV_{TT}.

aorta can be obtained noninvasively with a good degree of accuracy.

PWV estimates according to a new method based on area and flow (QA method (PWV_{QA})) were compared with PWV estimates based on wave transit time across the aortic arch (TT method (PWV_{TT})) from the same MRI acquisition for the same subjects (5). No statistical difference could be found, which further validates the QA method (Fig. 5). Both methods used here have the advantages of being noninvasive and applicable to the proximal aorta. The QA method allows for all measurements to be performed in the ascending aorta, and yields local PWV as well as compliance. On the other hand, the precise measurement of the area needed to calculate PWV_{QA} is not obtainable from the FLASH sequence acquired for the flow. With the sequences available on our scanner, the measurement of small area variations needs a second, specially-optimized acquisition sequence for the delineation of the aortic wall. However, an optimized segmented FLASH sequence providing accurate flow and area during the same acquisition certainly could be designed.

The accuracy and repeatability of the TT and BR methods are not evaluated in the literature against gold standards, such as pressure catheter measurements. However, in vivo animal as well as human studies (17,18) and in vitro studies on latex tubes (19) give a good assessment of the accuracy of MR flow measurement based on phase-contrast techniques.

In the TT method, the major source of error lies in the extrapolation of the steep upstroke of the curve to the baseline. The flow curves are much smoother than the area curves because the blood velocity is low near the arterial wall, and the total flow is therefore only slightly dependent on the sharpness of the aortic outline. This flow-based method applies to the entire aortic arch, which is curved and tapered. Therefore, deriving compliance from PWV is difficult, because it requires knowledge of the cross-sectional area. This means that it is necessary to use an appropriate mean area value in Eq. [7] to derive a compliance value from this PWV_{TT}.

A very simple and direct method that is often used to derive volume compliance or area compliance and PWV is the one based on area variation and pulse pressure (1,5). If applied to the aorta, the major problem lies in the pressure measurements, which have to be invasive to be direct. Noninvasive methods rely on brachial pulse pressure as a surrogate of central aortic pulse pressure; however, this is known to be a fairly imprecise approximation (20). Moreover, brachial pulse pressure is a better approximation of central pressure in the elderly than in young or middle-aged subjects, the latter being the group most concerned with an early detection of cardiovascular changes. The comparison of our new flow-area method (PWV_{QA}) with that based on brachial pulse pressure (PWV_{BR}) (Fig. 4) showed that the latter led to significantly higher PWV and consequently to lower compliance values. Due to increasing wall stiffness along the arterial tree, as well as to wave reflections, brachial pulse pressure is higher than central pulse pressure, thus leading to the discrepancies mentioned above.

Large-scale and long-term cardiovascular follow-up studies have provided evidence that pulse pressure, measured sphygmomanometrically at the brachial artery for practical reasons, is a good predictor of coronary heart disease (8). From a hemodynamical point of view, however, it is clear that the relevant pulse pressure influencing cardiac load is the central pulse pressure in the proximal aorta. Since brachial pulse pressure does not correspond to aortic pulse pressure, we could probably find more accurate risk factor indexes for obtaining central pulse pressure. This would indeed make it possible to detect patients with both a high risk of vascular modification of the proximal aorta and a fairly normal brachial pulse pressure. Moreover, measurement of the central pulse pressure is necessary for computation using the pulse pressure method, which has been shown to yield accurate estimates of total systemic compliance (21). The follow-up of central hemodynamic parameters might also increase in use, since its dependence on certain drugs makes it a new therapeutical marker (22–24).

With an adequately designed sequence, it should be possible to obtain both flow and area in the ascending aorta during the same acquisition, leading to a more accurate determination of the central PWV with the QA method than with both either of the other two methods.

To conclude, our new noninvasive method to derive aortic compliance based on MRI techniques proved to be accurate and in good agreement with a simple foot-to-foot transit time method, as well as with previously published results. Our results also suggest that brachial pulse pressure measurement is an inadequate way to obtain central compliance (in young and middle-aged subjects), since it overestimates the stiffness of the aorta. To get accurate estimates, central vascular properties must be derived from central measurements. These properties may in the future become increasingly useful as cardiovascular risk indices, while their direct estimates, obtained by noninvasive methods such as the one proposed here, may prove to be preferable for clinical use.

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