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Echocardiographic Assessment of Large Artery Stiffness

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Arterial wall stiffness is a highly relevant phenotype for cardiovascular medicine. Large artery stiffening is a key element in the pathogenesis of isolated systolic hypertension,¹ a condition mainly affecting older adults, which is responsible for a high burden of cardiovascular disease worldwide.² Arterial stiffening impairs the ability of conduit arteries to accommodate the stroke volume intermittently delivered by the left ventricle during systole, thus increasing the pulsatile hydraulic load impeding left ventricular ejection.^{3,4} There is increasing recognition of the adverse influence of increased pulsatile arterial load in conditions such as heart failure with preserved ejection fraction, heart failure with reduced ejection fraction, hypertensive heart disease, and valvular heart disease (particularly aortic stenosis). In addition, given the effect of arterial stiffness on pulsatile hemodynamics, increased large artery stiffness can also promote excessive penetration of pressure and flow pulsatility into the kidney and the brain, leading to target organ damage.⁵

Arterial stiffness is affected by various risk factors and biologic processes that lead to cardiovascular disease. Arterial stiffness increases with aging and various disease states, such as hypertension, diabetes mellitus, obesity, smoking, hypercholesterolemia, subclinical inflammation,⁶ and kidney disease.⁷ Various long-term exposures lead to an accelerated stiffening of large arteries with aging over a period of decades.⁸ Therefore, measurements of arterial stiffness not only provide information about the effects of prevalent conditions but also reflect the cumulative history of exposure to risk factors over a period of decades, much like hemoglobin A_{1c} levels reflect the cumulative exposure to hyperglycemia over a period of months. Unlike hemoglobin A_{1c}, however, arterial stiffness is not only a “marker” of upstream abnormalities but also a causal mediator of various forms of cardiovascular disease. Not surprisingly, available studies demonstrate that carotid-femoral pulse-wave velocity (PWV), an index of large artery stiffness, independently predicts the risk for incident cardiovascular events in both clinical and community-based cohorts.⁹⁻¹²

A precise in vivo characterization of the material properties of the arterial wall is not possible at present. However, arterial wall stiffness can be inferred or approximated using several indices derived from in vivo measurements of cyclic changes in pressure, diameter, and wall thickness at a given arterial location. These approaches yield intuitive metrics, such as cross-sectional compliance or distensibility coefficients.^{13,14} However, surface ultrasound can provide precise measurements of cyclic geometric changes only in superficial arteries, while ultrasound assessments of such changes in deeper, large arteries (such as the aorta),

have limited accuracy. Furthermore, the estimation of arterial compliance or distensibility from cyclic arterial geometric changes also requires local arterial pulse pressure measurements. Unfortunately, differences in pulsatile pressure profiles along the arterial tree lead to systematic and variable bias in measurements of large artery distensibility or compliance, when only standard brachial blood pressure measurements are used.

Measurements of arterial PWV, in contrast, do not require knowledge of local pressure-volume, pressure-diameter, or pressure-strain relationships. PWV is a functional parameter affected by the stiffness of the arterial wall. The propagation velocity of a pulse wave in an elastic tube can be described by the Bramwell-Hill equation, which relates PWV to the inverse of the cross-sectional distensibility coefficient (which is the fractional area change divided by the pressure change):

$$PWV = \sqrt{\frac{1}{\rho D_{\text{coeff}}}},$$

where ρ is the density of the fluid. It can be seen that PWV is related to the square root of the distensibility coefficient, which is in turn linearly related to both the elastic modulus of the wall material and to the arterial wall thickness/lumen ratio. The relationships between PWV, the incremental elastic modulus (E_{inc}) of the wall material, and arterial geometry, is best demonstrated by the Moens-Korteweg equation:

$$PWV = \sqrt{\frac{h E_{\text{inc}}}{D \rho}},$$

where h is wall thickness, D is vessel diameter, and ρ is the density of the fluid (which for in vivo measurements, represents blood density).

Perhaps the most attractive aspect of PWV as an index of arterial stiffness is the simplicity of its measurement. In principle, all that is required is a recording of the time-resolved pressure or flow profile at two arterial locations, as long as the distance (i.e., path length) between those locations is known. By measuring the pulse transit time (the time needed for the wave to travel from one measurement point to the other), PWV can be computed as path length/transit time. The timing of the onset of the pulse wave can be measured using either pressure or flow measurements, because both are manifestations of the same disturbance (i.e., the pulsatile energy traveling through the arterial wall).^{13,14} Transit time is typically measured by determining the time interval between the foot (takeoff) of the waveform at a proximal location and the foot of the waveform at a distal location. The reason for choosing the feet of the waveforms as landmarks is that the later portions of the waveforms are contaminated by reflected waves, whereas reflections do not prominently affect the early portions of the waveforms. Large artery PWV is most commonly measured over the carotid-femoral path using pressure measurements in these locations via applanation tonometry (Figure 1). Carotid-femoral PWV measured with arterial tonometry is a robust independent

predictor of cardiovascular risk.^{7,9-12} However, tonometric carotid-femoral PWV measurements are highly operator dependent and rely on the availability of dedicated PWV devices. Furthermore, some clinical conditions, like severe aortic stenosis or severe obesity, make carotid-femoral tonometry very difficult or impossible.

In this issue of *JASE*, Styczynski *et al.*¹⁵ report the results of a prospective study in which noninvasive Doppler ultrasound-based measurements of PWV between the aortic arch and the external iliac artery were compared with invasive catheter-based measurements. In this study, 45 subjects underwent simultaneous catheter-based and ultrasound-based measurements during a clinically indicated cardiac catheterization procedure. Transit times were measured from sequential aortic and external iliac pulsed-wave Doppler ultrasound recordings with electronic calipers, using the electrocardiographic S wave as a fiducial point. Analogous measurements were made from catheter-derived pressure recordings. The path length was measured using the intra-arterial catheters for invasive measurements, whereas surface measurements were used for ultrasound-based PWV. The investigators found a strong correlation between invasive and ultrasound-derived PWV (Spearman $R = 0.93$), with little systematic bias (~ 0.13 m/sec), adequate agreement between the methods (SD between measurements of ~ 0.79 m/sec), and low interobserver variability (intraclass correlation coefficient = 0.96, absolute difference between repeated measurements expressed as a percentage of the mean = $5.20 \pm 3.24\%$). On average, it took an experienced sonographer about 2 min to perform all recordings necessary for PWV measurements. The investigators appropriately conclude that wider implementation of echocardiography-derived PWV for evaluation of aortic wall stiffness can further expand the clinical and scientific utility of echocardiography.

Echocardiography remains the most widely available cardiovascular imaging technique in clinical practice. The use of this technique for PWV measurements is therefore not only an attractive means to expand the potential of echocardiography, but also should be of great interest to the field of arterial stiffness, because it would allow broader-scale assessments of aortic PWV in multicenter studies and clinical trials, and has great potential to accelerate the consideration of this phenotype in clinical practice.

Interestingly, echocardiography not only represents a suitable alternative for arterial tonometry to measure large artery PWV but also may have significant advantages over arterial tonometry for this particular purpose. For instance, ultrasound can access deep arterial locations (such as the aortic root, the abdominal aorta, and the descending thoracic aorta) that are not accessible by arterial tonometry (Figure 1). The ability to interrogate these locations, in turn, allows echocardiography to assess arterial segments that are particularly informative. For instance, transit times can be readily measured with Doppler ultrasound between the subclavian artery (interrogated from the suprasternal view) and the proximal abdominal aorta (interrogated from the subcostal view). This segment represents a “pure” elastic segment (unlike carotid-femoral PWV), along which noninvasive path length measurements are likely less subject to error, given that (1) this segment tends to be less tortuous than the abdominal aortic segment, which is included in the carotid-femoral path, and (2) the use of two-dimensional imaging can provide the distance from the suprasternal probe to the pulsed-wave Doppler sample volume (point of interrogation), which can in turn

be subtracted from the surface distance between the suprasternal notch and the distal point of interrogation, during computations of the arterial path length. Echocardiography can also provide the precise timing of the pulse onset at the left ventricular (LV) outflow tract, allowing the assessment of ascending aortic PWV, a segment that is entirely “missed” by tonometric carotid-femoral measurements, but is particularly important given its susceptibility to early arterial aging¹⁶ and its key role in ventricular-arterial interactions.^{3,4,17,18} In addition, LV outflow tract and proximal abdominal aortic interrogations can be used to measure the averaged PWV across the entire thoracic aorta. The thoracic aortic path length may potentially be derived from surface measurements of sternal length or body height. The latter approach has been recently used in children,¹⁹ but similar algorithms need to be developed in adults. Figure 1 summarizes various key arterial locations that can be interrogated for assessments of segmental large-artery PWV with ultrasound and arterial tonometry. Despite the great potential of echocardiography to interrogate key aortic segments, more research is needed to develop and validate algorithms to estimate path lengths of these segments on the basis of a combination of surface and on the basis of a combination of surface and 2D-ultrasound-derived distances. In the meantime, approaches such as the one described by Styczynski *et al.*¹⁵ or the approach previously described by Bonapace *et al.*²⁰ (which was shown to predict adverse outcomes in heart failure with reduced ejection fraction) represent readily implementable techniques that can add important information and minimal additional scanning time during transthoracic echocardiographic examinations.

Regardless of the arterial segment interrogated, several technical considerations are required when measuring pulse transit times with Doppler ultrasound. High sweep speeds (~150–200 mm/sec) are advantageous and provide enhanced precision to detect small time delays, particularly in stiff aortas with high PWV. A clear electrocardiographic signal with identical electrode placement and lead settings should be used for proximal and distal sequential recordings. Low-velocity filters (“wall filters”) should be eliminated to maximize the ability to record the earliest pulsatile flow (true foot) in each location. Similarly, the method to assess the timing of the foot of the flow wave is not trivial. Although Styczynski *et al.*¹⁵ used a simple visual assessment of the foot with caliper-based time measurements to maximize the simplicity of the method, this approach may introduce significant limitations in special circumstances, such as instances in which there is floor noise in flow measurements, significant flow reversal in end-diastole, or when time delays are particularly short (because of extremely high PWV values or relatively short arterial segments).

Perhaps the most important limitation of Doppler-based assessments of PWV is the effect of vortical or helical aortic flow on measurements of pulse arrival timing. Whereas the formation of small peripheral vortices in the thoracic aorta is a normal phenomenon, the presence helical flow or prominent, merged vortical flow tends to occur in various disease states (such as aortic dilation, aortic stenosis, bicuspid aortic valve, or among patients with a history of stroke) and leads to different flow velocity profiles (and timing) at different locations within the arterial lumen cross-section. This phenomenon can markedly confound assessments of the true timing of pulse arrival at a given point along the arterial path. Interestingly, this is not a concern with pressure-based assessments of PWV, because (for practical purposes) a single pressure waveform is present at any given point of the path

length, regardless of the presence of vortical flow, whereas multiple and markedly different flow velocity waveforms can be demonstrated within a single cross-section of the aorta (i.e., at a single level along the path length), when prominent helical or vortical flow occurs. A high degree of suspicion for the presence of aberrant flow patterns should be exercised, particularly in the presence of aortopathy or aortic valve abnormalities. Careful visual assessments of color Doppler images (Figure 2) and, when necessary, iterative pulsed-wave Doppler interrogations of flow velocities at different points in the lumen (i.e., slightly changing imaging planes and sample volumes from the centerline to the aortic wall) can be used to detect varying flow velocity profiles within the cross-section of the aortic lumen. When such phenomenon is present, Doppler interrogations cannot be relied upon for the assessment of the timing of pulse arrival. Importantly, helical and vortical flow patterns are particularly prevalent in the ascending aorta; therefore, the LV outflow tract (rather than the ascending aorta itself) may represent a more suitable location for assessing the proximal timing of the onset of flow, when the ascending aorta needs to be included in the segment being interrogated (Figure 1). Despite the excellent correlation and average agreement between echocardiography-based PWV and invasive PWV in the study by Styczynski *et al.*,¹⁵ it is likely that a combination of various sources of error mentioned above is responsible for the important differences (up to ~2 m/sec) between the 2 measurements observed in a minority of subjects in this study.

Although already an acceptable method for clinical use, there is plenty of “room” for improvement in echocardiography-based PWV measurements. In particular, the development of automated algorithms for flow envelope tracing, foot detection, electrocardiographic waveform extraction, and feature detection can yield significant improvements in the accuracy and precision of echocardiography-based PWV. Automated techniques for transit time estimations are well developed for tonometry-based assessments and have been applied to magnetic resonance imaging phase-contrast-based flow measurements but are not yet widely available for ultrasound-based flow velocity assessments. In our laboratory, we use MATLAB-based custom software to automatically trace ultrasound Doppler flow velocity envelopes from Digital Imaging and Communications in Medicine (DICOM) images, perform a color-based automated extraction of the electrocardiographic signal from these images, and compute arrival times on the basis of various automated methods (including regression-based projections of measured velocities toward the baseline, which is particularly useful in the presence of floor noise or excessive wall filtering that prevents direct visualization of the flow waveform foot). The software also processes the electrocardiographic signals on the basis of the entire morphology of the QRS complex (via cross-correlation) rather than using a single characteristic point (such as the S or R wave) as a fiducial point, thus enhancing the accuracy of transit time estimations between proximal and distal measurement points. Although image and signal processing is relatively complex in the “background,” such automated software algorithms could be easily implemented for fast online or offline measurements in the clinical work flow, enhancing the quality and reducing the time for echocardiography-based PWV measurements.

Given the growing appreciation for the importance of arterial loading on the left ventricle, it seems logical that echocardiography-based PWV measurements will aid in the clinical

phenotyping of patients with heart failure, hypertensive heart disease, aortic stenosis, and other valvular disorders. Importantly, notwithstanding the importance of PWV, transthoracic echocardiography provides even more exciting opportunities for phenotyping the hemodynamic function of the arterial tree in the era of precision medicine. In particular, comprehensive assessments of ventricular-arterial interactions and the hydraulic load (input impedance) that the arterial tree imposes on the left ventricle can be obtained with analyses of central arterial pressure-flow relations.¹⁸ Unlike effective arterial “elastance,” a commonly used index of arterial load that is essentially the product of microvascular resistance and heart rate (thus being insensitive to pulsatile arterial load or arterial stiffness),²¹ the proper assessment of pulsatile pressure-flow relations provides detailed and unique information about several important aspects of the ventricular-arterial cross-talk. This approach can be used to characterize the systolic load imposed by arterial wave reflections (which promotes LV remodeling and diastolic dysfunction and strongly predicts the risk for incident heart failure in the general population).^{17,22} In addition, this approach quantifies the load imposed by the aortic root and the total compliance of the arterial tree and, when combined with cardiac geometric measurements, can also characterize ejection-phase time-resolved wall stress patterns.^{3,4,17} These techniques have been applied to large cohort studies (such as the Asklepios study and the Framingham Heart Study), have yielded important insights into the effects of aging and various disease states on the arterial tree, and have characterized the physiologic effects of various vasoactive drugs in hypertension and heart failure.^{23,24} Their use in clinical research holds great potential. Importantly, analyses of pressure-flow relations require knowledge of the time-resolved central aortic pressure profile, in addition to Doppler flow profiles. Although central pressure assessments via carotid applanation tonometry are preferable for research-oriented measurements, carotid tonometry is highly operator dependent and unfeasible for high-throughput evaluations in multicenter studies or clinical practice. We are currently assessing the potential of operator-independent approaches suitable for analyses of pressure-flow relations in the echo laboratory, on the basis of either cuff-based brachial or wristwatch-based radial pressure waveform recordings (both coupled with mathematical functions in the frequency domain used to estimate the central pressure profiles).

Finally, advances in ultrasound technology will enable novel techniques for the assessment of arterial stiffness. For instance, although the foot-to-foot PWV technique using the heart-generated pulse wave is suitable for relatively long arterial segments, it is unsuitable for PWV assessments over very short segments, because of the minimal time delay between adjacent spatial points which are difficult to discern at the heart frequency. New techniques that measure the propagation of artificial waves in the arterial wall (such as those generated by tone bursts of focused ultrasound, whose speed of propagation can then be tracked with high temporal and spatial resolution) are being tested,²⁵ but their use will probably be limited to superficial arteries (such as the carotid).

To conclude, the contribution by Styczynski *et al.*¹⁵ is a significant step forward and should promote the application of echocardiography to phenotype the arterial system, allowing a more widespread application of the study of arterial stiffness and arterial load, not only in mechanistic, clinical and epidemiologic research but also in the translation of this important phenotype to clinical practice.

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Figure 1.

Potential arterial points of interrogation for measurements of large artery PWV. Five easily accessible arterial points for pulsed-wave Doppler flow velocity interrogations are shown (two of which are also accessible by arterial tonometry). These points can be used ad hoc to interrogate PWV in various segments. Points 1 and 3 can yield an assessment of ascending aortic PWV, an important aortic segment susceptible to early arterial aging, that is inaccessible by arterial tonometry. Interrogation of points 1 and 4 is highly feasible from standard echocardiographic views, but better algorithms to estimate thoracic path length from body height or body surface measurements are needed in adults. The use of points 3 and 4 is highly advantageous because the path length can be readily calculated on the basis of a combination of surface measurements and two-dimensional ultrasound-based measurements. However, this approach may be limited in patients with particularly stiff aortas (i.e., very high PWVs) because of the relatively short distance between the interrogation points, leading to significant measurement error. Strategies that use points 2 and/or 4 (including carotid-femoral PWV or aortic-femoral PWV) inevitably include long muscular arterial segments, which confound the assessment of large artery (aortic) stiffness, the phenotype of primary interest. The ability to exclude these muscular segments represents an advantage of echocardiography over arterial tonometry. Note that although muscular and elastic segments are sharply delimited in the figure, in reality, transitions between elastic and muscular arteries are gradual.

Figure 2.

Examples of aberrant flow in the thoracic aorta. Color Doppler imaging demonstrates varying flow velocities at different locations within the cross-section of the aortic lumen, at any given location of the aortic path length. In this case, marked vortical flow results in simultaneous forward and backward flow in the ascending aorta and aortic arch, demonstrated by color Doppler imaging. In this situation, the assessment of pulse arrival time from pulsed-wave Doppler interrogations is invalid.