P

atients with repaired coarctation of the aorta (CoA) are thought to have increased afterload due to abnormalities in vessel structure and function. However, blood pressure, which is the simplest measure of afterload, is not universally elevated in these patients. This may simply be because of inaccuracies introduced when measuring blood pressure brachially rather than centrally. It is also possible that blood pressure itself is a limited measure of vascular and cardiac load after CoA repair. In particular, it may not adequately characterize arterial wave reflections that may be more prevalent in this group.

Characterization of central aortic hemodynamics and wave reflections is difficult, requiring simultaneous measurement of aortic pressure and flow. We have recently demonstrated the ability to assess central aortic systolic blood pressure (c-SBP) using a combination of phase-contrast magnetic resonance and oscillometric brachial artery blood pressure.1 We have also shown that it is possible to use the same high temporal-resolution phase-contrast magnetic resonance data to perform noninvasive wave intensity analysis (WIA).2,3 This allows the assessment of wave reflections, which has not previously been done in the CoA population.

Cardiac MR (CMR) also provides accurate assessment of left ventricular (LV) hypertrophy, which is known to be a risk factor for adverse cardiac events. Thus, using CMR, it is possible to accurately assess all aspects of conduit vessel function and determine their relationship to end-organ effects.

In this study, we recruited 50 patients with repaired CoA (without recoarctation) and 25 age- and sex-matched healthy controls. The aims of the study were to use high temporal-resolution CMR imaging to (1) characterize differences in c-SBP and oscillometric brachial artery blood pressure.1 We have also shown that it is possible to use the same high temporal-resolution phase-contrast magnetic resonance data to perform noninvasive wave intensity analysis (WIA).2,3 This allows the assessment of wave reflections, which has not previously been done in the CoA population.

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DOI: 10.1161/HYPERTENSIONAHA.116.08763

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and peripheral systolic blood pressure (p-SBP) between patients and controls, (2) comprehensively evaluate afterload (including wave reflections) in the 2 groups, (3) identify possible biomarkers among covariates associated with elevated LV mass (LVM), and (4) substantiate in vivo observations with 1-dimensional (1D) computer simulations of wave travel in repaired coarctation.

Materials and Methods

Subjects
Fifty patients with CoA repaired in childhood and 25 healthy controls were recruited. Control subjects were volunteers from the same geographic area (Greater London) and were matched by age (within 2 years) and sex. Exclusion criteria were (1) irregular heart rates, (2) contraindications to CMR such as MR-incompatible implants, (3) pregnancy, (4) aortic stenosis, (5) coarctation associated with major or unrepaired congenital heart disease (exception nonstenotic bicuspid aortic valve or repaired ventricular/atrial septal defects), (6) coarctation stents, or (7) echocardiographic or CMR evidence of recoarctation (diastolic flow continuation in descending aorta or coarctation index [CI] <0.7). Patients receiving antihypertensive medications were included if this was reported as stable, chronic therapy. The study was performed with local research ethics committee approval and written informed consent was obtained.

CMR Protocol
All imaging was performed on a 1.5-T MR scanner (Avanto, Siemens Medical Solutions, Erlangen, Germany) using 2 spine coils and 1 body-matrix coil. A vector electrocardiographic system was used for cardiac gating. The flow-imaging plane was planned using orthogonal long axis cine images of the ascending aorta and was placed just above the sinotubular junction (Figure S1, Line A). These data were used for the derivation of all subsequent hemodynamic indices. The sequence was a prospectively triggered, spiral, velocity-encoded spoiled gradient echo acquisition, accelerated with SENSE (TE/TR: 1.9/4.8 ms, FOV: 400x400x66 mm, matrix: 192x192, spiral interleaves: 60, SENSE factor: 4, VENC: 180 cm/s). The temporal resolution was 9.6 ms, the spatial resolution was 2.1x2.1 mm, and the breath-hold was ±11 s (16 R–R intervals).

Aortic arch morphology was assessed in patients using gadolinium-enhanced MR angiography with a coronal 3D fast-field-echo sequence or a balanced, steady-state free precession sequence. A vector k-space SENSE sequence was used to calculate LV volumes and LVM as previously described (online-only Data Supplement).

Blood Pressure Measurement
Brachial SBP (p-SBP), diastolic BP (DBP), and mean BP were measured by automated oscillometric sphygmomanometry during flow imaging (Datex Ohmeda) on the patient’s right arm. Small-adult, adult, and large-adult cuff sizes were chosen according to subject arm circumference. Volunteers lay supine in the CMR scanner with the arm at the level of the heart, and there was a period of acclimatization (at least 15 minutes) before measurements were taken.

Image Processing
All images were processed using an in-house plug-in for the open source DICOM software OsiriX (OsiriX Foundation, Geneva, Switzerland). Segmentation of the ascending aorta was performed on the modulus image using a previously validated semiautomatic registration-based algorithm (online-only Data Supplement).

Aortic arch anatomy was evaluated by measurement of aortic diameter in the transverse aortic arch between innominate and left common carotid artery, distal aortic arch (repair site), and the descending aorta at the level of the diaphragm (Figure S1). Two metrics were derived from these measurements: (1) transverse arch index (TI), quantifying the degree of transverse arch hypoplasia: transverse arch diameter divided by descending aorta diameter. (2) CI, quantifying the degree of recoarctation: aortic isthmus (repair site) diameter divided by descending aorta diameter. Aortic arches were also characterized as “gothic” if the arch had an acutely angulated triangular configuration.

Derivation of c-SBP Using Area-Distension Waveforms
The aortic area waveforms were calibrated using a previously validated exponential pressure–area model and described below. The equation of an exponential pressure–area relationship is: 11,12

\[ p(t) = p_s \exp\left(\frac{A(t)}{A_d}\right) \]

where \( p(t) \) is the synthesized pressure curve, \( p_s \) is the brachial DBP, \( A_d \) is the diastolic area, \( A(t) \) is the area curve, and \( \beta \) is the scaling factor, initially set as follows:

\[ \beta = \frac{A_d \ln\left(\frac{p_s}{p_d}\right)}{A_d - A_s} \]

where \( A_s \) and \( A_d \) are the systolic and diastolic aortic areas, respectively. As c-SBP is, in general, lower than p-SBP, this initial starting \( \beta \) is the theoretical maximum.

Calibration of this model consisted of iteratively reducing the scaling factor \( \beta \) to minimize the difference between the measured brachial mean BP and the mean of the synthesized pressure curve. This calibration scheme was based on the validated assumption that the difference between DBP and mean pressure is constant in the large arteries.13,14 The estimated c-SBP was the peak of the optimized synthesized pressure curve.

Wave Speed and Characteristic Impedance
Because of the possible early reflection site related to the repaired coarctation, conventional single cut methods of calculating wave speed (such as the Q/A method) are unreliable.15,16 Therefore, the Bramwell–Hall equation17,18 was used to obtain local pulse wave velocity, \( c \), in the ascending aorta, using the aortic area waveform and c-SBP:

\[ c = \sqrt{\frac{A_c \Delta P}{\rho \Delta A}} \]

where \( A_c \) is the diastolic cross-section area, \( \Delta A = A_d - A_s \) and \( \Delta P \) is the central pulse pressure (c-SBP–DBP). This method has been shown in computer simulations to be robust in the presence of early wave reflections.15 Characteristic impedance, \( Z_c \), was calculated by:

\[ Z_c = \rho c / A_f \]

where \( \rho \) is assumed to be 1060 kg/m³.

Wave Intensity Analysis
In WIA, waves are regarded as a summation of incremental wave fronts; it is, therefore, possible to separate the \( Q/A \) method. In WIA, waves are regarded as a summation of incremental wave fronts; it is, therefore, possible to separate the

Arterial Resistance and Total Arterial Compliance
Arterial resistance (Woods Units) was calculated by dividing mean BP (mmHg) by cardiac output (L/min). Total arterial compliance

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(TAC, mL mmHg⁻¹) was calculated in Matlab using a 2-element windkessel model as previously described, using single cardiac cycle phase-contrast flow curves and both central (TACcentral) and brachial (TACbrachial) pulse pressure (SBP–DBP). Briefly, the aortic flow curve was inputted into the model with measured arterial resistance, and TAC was tuned so that pulse pressure generated by the model equaled measured pulse pressure. Resistance was indexed to body size by multiplication with BSA and TAC by dividing by BSA.

One-Dimensional Computer Simulations
A validated 1D model of the systemic vascular tree was used to explore differences in wave reflection between controls and patients after coarctation repair. The model solves the 1D Navier–Stokes equations and provides pressure and flow waveforms along the arterial tree. The model was run using a time-varying elastance model for the heart on its upstream boundary, such that simulated waves originate from the interaction of the pumping heart in the arterial tree. Apart from the baseline condition using default parameters for the model (default lengths, inlet and outlet diameters and arterial distensibility), 2 additional scenarios were simulated. First, the diameter of a segment in the descending aorta was reduced by 25% to mimic a mild residual aortic coarctation, while maintaining the normal distensibility. Second, the stiffness of the narrowed section was increased by 5 orders of magnitude to mimic the effect of a stiffened coarctation repair zone.

Statistics
STATA 13.1 and Graphpad Prism 5 were used for statistical analysis and figures. Data were examined for normality and where appropriate, non-normally distributed variables were log transformed to ensure normal distribution before analysis. Descriptive statistics are expressed as mean (±95% confidence interval) when normally distributed, and geometric mean (±95% confidence interval of geometric mean) when non-normally distributed, unless specified. Proportions are expressed as percentages.

The independent samples t test was used to compare differences in parametric data between coarctation patients and controls; Welch correction was used for unequal variances. Proportions test was used to compare proportions among groups. The level of α considered for statistical significance was 0.05.

Pearson correlation coefficient was used to analyze simple linear relationships between measures of arch morphology and hemodynamics. Multivariable linear regression analysis was also used to determine covariates independently associated LVM. The model was adjusted for: BSA, sex, age to ascertain independent associations. We also adjusted for case/control to control for unmodeled covariates.

Results

Study Population Characteristics
Demographics and conventional clinical measures in the patients and controls are shown in Table 1. There were no significant differences in age, sex, or ethnicity between patients and controls. Control subjects were taller on average than patients (P=0.03); however, there was no difference in BMI or BSA between groups (Table 1).

The clinical characteristics of the coarctation patients are shown in Table S1. The median age at repair was 4.0 months (interquartile range 0.5–48.2 months). Eighteen of fifty patients (36%) underwent repair in the neonatal period and 34 of 50

Table 1. Study Population Demographics and Basic Hemodynamics

<table>
<thead>
<tr>
<th>Parameters</th>
<th>Patient, n=50</th>
<th>Control, n=25</th>
<th>Significance</th>
</tr>
</thead>
<tbody>
<tr>
<td>Male (%)</td>
<td>35 (70)</td>
<td>18 (72)</td>
<td>P=0.9</td>
</tr>
<tr>
<td>White/Asian (%)</td>
<td>43 (86)/7 (14)</td>
<td>23 (92)/2 (8)</td>
<td>P=0.7</td>
</tr>
<tr>
<td>Age, y.†</td>
<td>24 (22–27)</td>
<td>23 (21–25)</td>
<td>P=0.3</td>
</tr>
<tr>
<td>Weight, kg*</td>
<td>71 (67–76)</td>
<td>76 (69–83)</td>
<td>P=0.3</td>
</tr>
<tr>
<td>Height, cm*</td>
<td>171 (168–174)</td>
<td>177 (173–180)</td>
<td>P=0.03</td>
</tr>
<tr>
<td>Body surface area, m²</td>
<td>1.8 (1.8–1.9)</td>
<td>1.9 (1.8–2)</td>
<td>P=0.1</td>
</tr>
<tr>
<td>Body mass index, kg/m²</td>
<td>24 (23–26)</td>
<td>24 (22–26)</td>
<td>P=0.9</td>
</tr>
<tr>
<td>Brachial systolic BP, mm Hg</td>
<td>122 (118–125)</td>
<td>117 (112–121)</td>
<td>P=0.1</td>
</tr>
<tr>
<td>Brachial mean BP, mm Hg</td>
<td>87 (85–89)</td>
<td>85 (82–89)</td>
<td>P=0.4</td>
</tr>
<tr>
<td>Brachial diastolic BP, mm Hg</td>
<td>63 (61–66)</td>
<td>64 (61–68)</td>
<td>P=0.6</td>
</tr>
<tr>
<td>Heart rate, bpm</td>
<td>67 (64–70)</td>
<td>64 (59–68)</td>
<td>P=0.2</td>
</tr>
<tr>
<td>LV ejection fraction, %†</td>
<td>65 (62–67)</td>
<td>63 (61–65)</td>
<td>P=0.2</td>
</tr>
<tr>
<td>LV mass index, g/m²</td>
<td>72 (68–76)</td>
<td>59 (55–63)</td>
<td>P&lt;0.0005</td>
</tr>
<tr>
<td>Cardiac index, L/min per m²†</td>
<td>3.1 (2.9–3.3)</td>
<td>3.3 (3.1–3.5)</td>
<td>P=0.2</td>
</tr>
<tr>
<td>Resistance index, Wood units.m²†</td>
<td>28 (26–30)</td>
<td>26 (24–27)</td>
<td>P=0.07</td>
</tr>
<tr>
<td>TACm, mL/m² per mmHg*</td>
<td>0.52 (0.58–0.67)</td>
<td>0.62 (0.57–0.67)</td>
<td>P=0.0007</td>
</tr>
<tr>
<td>Aorta diastolic area, cm²†</td>
<td>5.5 (4.9–6.1)</td>
<td>4.8 (4.5–5.1)</td>
<td>P=0.2</td>
</tr>
<tr>
<td>Ascending-transverse arch index‡</td>
<td>0.73 (0.69–0.77)</td>
<td>0.86 (0.82–0.89)</td>
<td>P&lt;0.0002</td>
</tr>
<tr>
<td>Transverse arch index</td>
<td>1.0 (1.0–1.1)</td>
<td>1.2 (1.2–1.3)</td>
<td>P&lt;0.0005</td>
</tr>
<tr>
<td>Coarctation index‡</td>
<td>0.9 (0.9–1.0)</td>
<td>1.1 (1.1–1.2)</td>
<td>P&lt;0.0005</td>
</tr>
</tbody>
</table>

BP indicates blood pressure; LV, left ventricle; and TAC, total arterial compliance.
† t test on log-transformed data: geometric mean and (95% confidence interval of geometric mean).
‡ Welch correction for unequal variance.
(68%) patients had their repair <1 year of age. End-to-end anastomosis was performed in 34 of 50 (68%) patients, extended end-to-end anastomosis in 4 of 50 (8%) patients, subclavian flap angioplasty in 10 (20%) patients, and Dacron/Gore-Tex patch augmentation in 2 (4%) patients. A bicuspid aortic valve was present in 26 of 50 (52%) patients. Eighteen percent of patients were receiving stable antihypertensive therapy. A single agent was prescribed in 8 patients and a combination of agents in 2 patients (2 and 4 agents, respectively; Table S1).}

There were no cases of recoarctation on CMR, with all cases having a CI >0.7: group median CI was 0.91 (10th–90th percentile, 0.78–1.14). In keeping with this finding, no agents in 2 patients (2 and 4 agents, respectively; Table S1).

**Central Hemodynamics and Wave Intensity**

<table>
<thead>
<tr>
<th>Parameters</th>
<th>Patient, n=50</th>
<th>Control, n=25</th>
<th>Significance</th>
</tr>
</thead>
<tbody>
<tr>
<td>c-SBP, mmHg</td>
<td>113 (110–117)</td>
<td>107 (103–110)</td>
<td><em>P</em>&lt;0.01</td>
</tr>
<tr>
<td>TACcentral mL/m² per mmHg*</td>
<td>0.62 (0.58–0.67)</td>
<td>0.79 (0.72–0.86)</td>
<td><em>P</em>&lt;0.0004</td>
</tr>
<tr>
<td>Pulse wave velocity, m/s*</td>
<td>4.2 (3.9–4.6)</td>
<td>3.7 (3.4–4.1)</td>
<td><em>P</em>&lt;0.04</td>
</tr>
<tr>
<td>Stiffness index, β²,†</td>
<td>1.68 (1.45–1.95)</td>
<td>1.33 (1.10–1.60)</td>
<td><em>P</em>&lt;0.05</td>
</tr>
<tr>
<td>Characteristic impedance, dyn/s per cm²,†</td>
<td>82 (74–90)</td>
<td>83 (76–91)</td>
<td><em>P</em>&lt;0.8</td>
</tr>
<tr>
<td>BCW, cm²,†</td>
<td>0.001 (0.0008–0.0014)</td>
<td>0.0007 (0.0005–0.0009)</td>
<td><em>P</em>&lt;0.01</td>
</tr>
<tr>
<td>BCW arrival, ms</td>
<td>88 (80–97)</td>
<td>111 (99–123)</td>
<td><em>P</em>&lt;0.003</td>
</tr>
<tr>
<td>FCW, cm²,†</td>
<td>0.009 (0.007–0.01)</td>
<td>0.01 (0.009–0.01)</td>
<td><em>P</em>&lt;0.3</td>
</tr>
<tr>
<td>FEW, cm²</td>
<td>0.0017 (0.0015–0.0021)</td>
<td>0.0016 (0.0012–0.0021)</td>
<td><em>P</em>&lt;0.5</td>
</tr>
<tr>
<td>Reflection magnitude†</td>
<td>0.91 (0.90–0.92)</td>
<td>0.88 (0.87–0.89)</td>
<td><em>P</em>&lt;0.001</td>
</tr>
</tbody>
</table>

BCW indicates backward compression waves; c-SBP, central aortic systolic blood pressure; FCW, forward compression waves; FEW, forward expansion waves; and TAC, total arterial compliance.

*Welch correction for unequal variance.
There was no significant difference in these metrics in subjects with a gothic arch.

**Determinants of LVM**

There was a significant difference ($P<0.00005$) in LVM index between patients and controls, 72 g/m² (68–76 g/m²) versus 59 g/m² (55–63 g/m²). There was no difference in LVM index between patients with bicuspid and those with tricuspid aortic valves ($P=0.7$).

The determinants of LVM were assessed using univariable linear regression analysis. Models were adjusted for known predictors of LVM: BSA, age, and sex, associations shown in Table 3.

On multivariable regression analysis, including significant measures from univariable analysis, only BCW log had a significant independent association with LVM ($P=0.013$; Table 3).

There was no significant independent association of LVM with c-SBP, TI, or CI.

**Discussion**

In this study, we used advanced CMR techniques to simultaneously assess all the components of afterload (vascular resistance, TAC, pulse wave velocity, and wave reflections) in patients with repaired coarctation and no significant recoarctation. The main findings of the study were (1) central SBP was higher in patients compared with controls, although there was no significant difference in brachial SBP; (2) patients postcoarctation repair were characterized by lower TAC, higher local wave speed, and increased BCW, without differences in aortic root $Z_c$; and (3) higher LVM was observed in patients and was significantly associated with the magnitude of the BCW.
Arterial Stiffness

The simplest measure of afterload is SBP. However, assessment of SBP alone does not specify which components of afterload are abnormal. Therefore, we assessed total and local arterial stiffness, as well as resistance and cardiac output in patients with repaired coarctation. The main findings were reduced stiffness, as well as resistance and cardiac output in patients because of the opposing effects of wall stiffening and aortic root dilation. This emphasizes that the observed differences in pulsatile load are more related to abnormalities in the distal portions of the aorta (such as the repair site). A limitation of this study was that local vessel properties distal to the repair site were not assessed. Consequently, it is not known whether descending aortic stiffness is also increased and this is an important area to investigate in a future study.

Wave Intensity Analysis

The presence of abnormal wave reflection has not previously been assessed in patients with repaired coarctation. Therefore, the increased magnitude of the BCW in patients is an important and novel finding. In patients with significant recoarctation, anatomic impedance mismatching at the site of stenosis would result in wave reflection. In this study, patients with recoarctation were specifically excluded and there was no significant relationship between BCW magnitude and anatomic narrowing. Therefore, an alternative mechanism for the generation of BCWs is required. To investigate possible causes, we used a 1D computer model of the vasculature that can replicate basic pressure and flow physiology. In a model with a mild anatomic narrowing, a small additional BCW was produced. However, only after increasing the stiffness of the repair site did the magnitude of BCW reflect the in vivo findings. Stiffening of the repair site is consistent with fibrous scar, and our findings suggest that this is an important mechanism in the generation of wave reflections.

It should be noted that there were some important differences between the in vivo and 1D model results. For instance, in the modeled WIA data, there is an early backwards expansion wave that is not present in the in vivo data. However, as this phenomenon has not been demonstrated in any previous in vivo studies, we believe it is an artifact of the model itself. A more important difference is related to the morphology of the BCW. In the modeled data, patients seem to have 2 backwards compressions waves. The first is presumably related to reflections from the coarctation repair site, whereas the second later wave (also present in modeled controls) is likely the result of more peripheral vasculature reflections. In contrast, most patients have only one BCW, although it did arrive slightly earlier than in controls. One possible explanation for this disparity is the relatively low temporal resolution of phase-contrast magnetic resonance, which may cause summation of the 2 backwards waves present in most patients. This would result in the appearance of an earlier arriving, higher magnitude single BCW in patients compared with controls. Wave entrapment and vascular hori-

Table 3. Univariable and Multivariable Relationships of Covariates With LV Mass (g)

<table>
<thead>
<tr>
<th>Parameters</th>
<th>Univariable Analysis</th>
<th>Multivariable Analysis</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>β</td>
<td>Significance</td>
</tr>
<tr>
<td>BCW</td>
<td>0.30</td>
<td>P=0.001</td>
</tr>
<tr>
<td>c-SBP</td>
<td>0.2</td>
<td>P=0.03</td>
</tr>
<tr>
<td>p-SBP</td>
<td>0.06</td>
<td>P=0.5</td>
</tr>
<tr>
<td>TAC</td>
<td>−0.12</td>
<td>P=0.22</td>
</tr>
<tr>
<td>TI</td>
<td>−0.2</td>
<td>P=0.02</td>
</tr>
<tr>
<td>CI</td>
<td>−0.2</td>
<td>P=0.01</td>
</tr>
<tr>
<td>Case/Control</td>
<td>0.4</td>
<td>P=0.0005</td>
</tr>
</tbody>
</table>

Each row represents separate model adjusted for BSA, age, and sex. BCW indicates backward compression waves; CI, coarctation index; c-SBP, central aortic systolic blood pressure; p-SBP, peripheral systolic blood pressure; TAC, total arterial compliance; and TI, transverse arch index.

*Log normalized variable.
arise more proximally in the vasculature—the vascular horizon. Because an earlier arriving wave would be more susceptible to summation, wave entrapment might help explain the usual single BCW seen in patients. An earlier arriving BCW could also explain the surprisingly small difference in BCW arrival times in patients versus controls. This could be investigated by performing noninvasive WIA distal to the repair site and would represent an important improvement for future studies. Nevertheless, our findings are consistent with previously published fluid structure interaction models of coarctation. These studies emphasize the role of stiffening as an important possible source of increased BCWs from the arch.\textsuperscript{25,26}

In this study, there were no significant differences in the forward compression or expansion waves between patients and controls. However, these waves can modify the morphology of the aortic pressures curve. For instance, the “proto-diastolic” forward expansion wave is thought to cause the late systolic inflections that occur in the absence of reflected waves. This was evident in our data (Figure 1), but was not different between the groups. Nevertheless, the magnitude of the forward expansion waves is thought to change with worsening diastolic function and in these patients may be of importance.

**Blood Pressure Differences**

We found no difference in brachial measured SBP between patients and controls. Although this is in keeping with previous studies, it does not mean that afterload is normal post coarctation repair. It is well recognized that brachial SBP overestimates central SBP,\textsuperscript{21,27} as seen in our data. This phenomenon is because of patient-specific pressure augmentation and explains the elevated central, but not brachial SBP in our patients. Importantly, a significant proportion of patients would be reclassified as hypertensive using c-SBP. These data suggest that reliance on brachial blood pressure may potentially result in undertreatment of at-risk patients. Failure to address blood pressure adequately in this young population would be reclassified as hypertensive using c-SBP. This was evident in our data (Figure 1), but was not different between the groups. Nevertheless, the magnitude of the forward expansion waves is thought to change with worsening diastolic function and in these patients may be of importance.

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We found no difference in brachial measured SBP between patients and controls. Although this is in keeping with previous studies, it does not mean that afterload is normal post coarctation repair. It is well recognized that brachial SBP overestimates central SBP,\textsuperscript{21,27} as seen in our data. This phenomenon is because of patient-specific pressure augmentation and explains the elevated central, but not brachial SBP in our patients. Importantly, a significant proportion of patients would be reclassified as hypertensive using c-SBP. These data suggest that reliance on brachial blood pressure may potentially result in undertreatment of at-risk patients. Failure to address blood pressure adequately in this young population would be reclassified as hypertensive using c-SBP. This was evident in our data (Figure 1), but was not different between the groups. Nevertheless, the magnitude of the forward expansion waves is thought to change with worsening diastolic function and in these patients may be of importance.

**Left Ventricular Mass**

In this study, patients with repaired coarctation had elevated LVM, which has been shown to strongly predict increased cardiovascular morbidity and mortality.\textsuperscript{35,36} In multivariable analysis, the most important hemodynamic association with LVM was the magnitude of the BCW. This is in keeping with previous human and animal studies that have shown that wave reflections play a prominent role in determining LV structure and function. In the multiethnic study of atherosclerosis, backward waves have been shown to be associated with both LVM\textsuperscript{37} and heart failure events.\textsuperscript{38} Furthermore, Hashimoto et al\textsuperscript{17} identified that antihypertensive therapy reduced backward waves, which may have an important role in mediating the ventricular response to therapy. The exact mechanism by which early BCWs contribute to the development of increased LVM is uncertain. However, it may be related to reflected waves altering the temporal evolution of LV pressure. For instance, Kobayashi et al\textsuperscript{40} showed a more profound hypertrophic response in rats banded in the descending aorta compared with the ascending aorta. This was thought to be because of late systolic loading due to wave reflections being a more potent hypertrophic stimulus. Irrespective of the mechanism, these data demonstrate that afterload cannot simply be thought of as SBP.

**Novel CMR Imaging**

Comprehensive noninvasive assessment of hemodynamics was made possible in this study because of the novel CMR and signal processing techniques used. All of the hemodynamic data were derived from a single high temporal-resolution phase-contrast acquisition, obtained during a breath-hold. These data provide the prerequisite area and flow data needed to derive central systolic pressure, resistance, TAC, local pulse wave velocity, and wave intensity when combined with simultaneous brachial noninvasive blood pressure measurement.

In this study, TAC was assessed using a technique based on 2-element Windkessel modeling. This technique is known to be an improvement on the stroke volume/pulse pressure method, which overestimates TAC. However, this technique has been criticized because it does not account for wave reflections. Nevertheless, it does provide and easily understood metric of arterial buffering that captures the simpler elements of pulsatile load. Consequently, we believe that it is a useful adjunct to the more sophisticated WIA performed in this study.

We also used the Bramwell–Hill equation for the derivation of local pulse wave velocity/characteristic impedance, rather than the flow-area (QA) or impedance-based methods. This was because we anticipated significant early reflections during the early part of systole, which would have resulted in errors in estimation of pulse wave velocity.\textsuperscript{15}

WIA provides useful insight into wave reflections in the time domain, it is relatively simple to implement but requires high temporal-resolution data flow imaging to resolve waveforms, which have short durations.
The ability to assess c-SBP, TACcentral, and characteristic impedance during routine CMR assessment is feasible, whereas performing noninvasive WIA may be more challenging to implement. However, the association with LVM suggests that these parameters may be useful biomarkers. Further studies assessing their relationship to outcomes and response to therapies are necessary.

Limitations
A significant limitation of this study is the possibility of cascading errors resulting from inaccuracies in the multiple acquisition and processing steps required to perform noninvasive WIA. These errors could originate from residual aliasing in the reconstructed images, incorrect segmentation, inaccurate blood pressure measurement, and inadequate assumptions. Nevertheless, the relationship between LVM and BCW does add credence to our findings.

Central SBP was derived by calibrating aortic area curves to mean and DBP using an exponential pressure–area relationship. This method was originally validated using noninvasive carotid tonometry (over a similar pressure range in this study), rather than using preferred invasive pressures. Unfortunately, micromanometer pressure catheters are not available for the magnetic resonance imaging environment and the alternative, fluid-filled catheters, have unfavorable frequency response and damping characteristics, which limit their use for this purpose. We have also previously described the use of a more complicated “arctangent” pressure–area model; however, this method cannot be reliably calibrated using noninvasive local pulse wave velocity (Q/A method), as it underestimates wave speed in the presence of significant early wave reflections, as observed in coarctation patients.15

The units of noninvasive WIA in this study, cm5 do not have an easily understandable physical meaning: in contrast to the W/m2 units of conventional WIA. Nevertheless, the waveforms produced by noninvasive WIA are qualitatively similar to invasive WIA in the literature—and given a linear pressure–area relationship would be directly proportional. Area and flow waves can, therefore, be considered analogous to pressure and velocity waves as found in the WIA literature.

Conclusions
This study aimed to describe all relevant central hemodynamic parameters underlying the vascular abnormalities in CoA. We have shown that this can be performed within a single breathhold, using a high temporal-resolution phase-contrast acquisition in the ascending aorta with simultaneous oscillometric blood pressure measurement. These data provide central blood pressure, systemic vascular resistance, TAC, local pulse wave velocity, and wave reflections.

We have shown that patients with repaired coarctation have higher c-SBP, stiffer aortas, and abnormal wave reflections, despite similar p-SBP to controls. Elevated LVM in this population is associated with the magnitude of the BCW. Consequently, therapies that can influence TAC and wave reflections may represent important drug development targets for the future.

Perspectives
Understanding the vascular abnormalities in patients with CoA is important if we wish to modify the cardiovascular risk and excess mortality in this population.41 Conventional clinical assessment using brachial blood pressure and anatomic imaging may fail to identify patients with vascular dysfunction, thereby leading to inappropriate undertreatment. Future therapeutic interventions should be targeted to the vascular abnormalities, which occur in this population.

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Disclosures
None.

References


Abnormal Wave Reflections and Left Ventricular Hypertrophy Late After Coarctation of the Aorta Repair

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Supplement

Abnormal Wave Reflections and LV Hypertrophy Late After Coarctation of the Aorta Repair

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Supplemental Methods

CMR Protocol
LV Volume and Mass
Images were acquired in the ventricular long-axis, four chamber, and the short axis covering both ventricles (9-12 slices). Assessment of the LV volume was performed by manual segmentation of the endocardial contour of short-axis cine images at end-diastole and end-systole using a built-in plug-in for OsiriX. End diastolic volume and end systolic volume were calculated using Simpson’s rule. From these volumes, stroke volume and ejection fraction (EF) were calculated. Epicardial contours were manually segmented at end systole. Ventricular mass was calculated as the difference between the epicardial and endocardial contours multiplied by the slice thickness and a specific gravity of ventricular mass of 1.05g/ml. LV mass was adjusted for body surface area (BSA) to provide the indexed LV mass (g/m²).

Aortic Arch Anatomy
Aortic arch morphology was assessed in patients using gadolinium-enhanced MR angiography as previously described with a coronal 3D fast-field-echo sequence. Gadolinium (Dotarem, Guerbet) was injected into a peripheral vein and tracked into the heart with a dynamic coronal 2D fast-field-echo sequence. The gadolinium dose was 0.2 mmol/kg. The MR angiographic sequence was started when contrast reached the left ventricle. Two consecutive angiograms were acquired in a single 15- to 20-second period of apnea. In control subjects and 7 patients intravenous contrast was not administered (IV access could not be obtained [n=2] or no consent for contrast [n=5]). Aortic anatomy was assessed in these subjects using a balanced, steady-state free precession sequence.

Image Processing
All images were processed using an in-house plug-in for the open source DICOM software OsiriX (OsiriX Foundation, Geneva, Switzerland). Segmentation of the ascending aorta was performed on the modulus image using a previously validated semi-automatic registration-based algorithm. The aortic region of interest (ROI) was manually adjusted as necessary to ensure optimal vessel wall delineation. The final ROIs were used to both calculate the aortic cross-sectional area (A) and prescribe the region in the phase image from which mean velocity (V) and flow (Q) were calculated. However, pressure wave synthesis requires curves that span the whole RR interval and the use of prospective gating resulted in missing data in the last 80-100ms of diastole. These missing data were recovered using linear interpolation between the last point and the first point. The curves were then linearly interpolated to 1ms temporal resolution and filtered using a zero-phase, low-pass, 2nd order Butterworth filter with normalized cut-off frequency of 0.04 (20Hz), (Matlab 2014b, Mathworks).

Aortic arch anatomy was evaluated by measurement of aortic diameter in the transverse aortic arch between innominate and left common carotid artery, distal aortic arch (repair site), and the descending aorta at the level of the diaphragm. Two metrics were derived from these measurements (Figure S1): (1) Transverse arch index (TI), quantifying the degree of transverse arch hypoplasia: transverse arch diameter divided by descending aorta diameter. (2) Coarctation index (CI), quantifying the degree of recoarctation: aortic isthmus (repair site) diameter divided by descending aorta diameter. Aortic
arcs were also characterised as ‘gothic’ if the arch had an acutely angulated triangular conformation.5

Wave Intensity Analysis
In WIA, waves are regarded as a summation of incremental wave fronts; it is therefore possible to separate the \( Q \) and \( A \) curves into the respective forward (+) and backward (-) components by expressing the relationship between wave speed, \( c \) (Equation 1) and changes in flow and cross sectional area.

\[
c = \pm \frac{dQ \pm}{dA \pm}
\]

Equation 1

Equation 1 combined with Equations 2 and 3:

\[
dA = dA_+ + dA_-
\]

Equation 2
\[
dQ = dQ_+ + dQ_-
\]

Equation 3
can be solved for the changes in the forward and backward flow and cross-sectional area; this results in Equations 4 and 5:

\[
dQ = \frac{1}{2}(dQ \pm c dA)
\]

Equation 4
\[
dA = \frac{1}{2}(A \pm \frac{1}{c} dQ)
\]

Equation 5

Net wave intensity \( dI_a \) was defined as the product of the differentials of cross-sectional area and flow.

\[
dI_a = dA \ dQ
\]

Equation 6

Similarly it can be shown that the net wave intensity \( dI_a \) (Equation 6) can be divided into the forward and backward intensities, Equation 7:

\[
dI_a = dI_{a(+) +} + dI_{a(-) -}
\]

Equation 7

with the separated \( dI_a \) expressed as:

\[
dI_{a(\pm)} = \pm c \left[ dA \pm \frac{dQ}{c} \right]^2
\]

Equation 8

Using this formulation and the value of \( c \) calculated using the Bramwell-Hill equation, forwards and backwards \( dI_a \) were calculated and plotted. As per convention, the direction of waves was referenced to the direction of blood flow. Waves arising from the heart were defined as forward running and those arising from the vasculature as backward running. Waves causing an increase in area were classified as compression waves and those causing a decrease in area as expansion waves by examination of \( dA_\pm \) plots. Thus, a forward running wave was held to be a compression wave if \( dA_+ \) was greater than zero and an expansion wave if \( dA_+ \) was less than zero. Similarly, a backward-running wave was considered as a compression wave if \( dA_- \) was greater than zero and an expansion wave if \( dA_- \) was less than zero.
Using this system four different waves may be characterised: Forwards Compression Waves (FCW), Forwards Expansion Waves (FEW), Backwards Compression Waves (BCW) and Backwards Expansion Waves (BEW).

The type of wave and their magnitude (area under the wave) were determined by analysis of the net and separated WIA plots in Matlab. The areas under the separated waveforms were calculated by numerical integration. Area waveforms were also separated into forward and backward components, by integration of dA+ and dA− plots. Using these data, we calculated the reflection magnitude as: \( \frac{\text{Area}_{\text{backward}}}{\text{Area}_{\text{forward}}} \).
Supplement References


<table>
<thead>
<tr>
<th>Type of Repair</th>
<th>Age of Repair</th>
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| End-to-End Anastomosis        | Neonatal Repair | 18 (36%)  
| Extended End-to-End Anastomosis | Repair <1 year | 34 (68%)  
| Subclavian Flap Angioplasty   | 34 (68%)      |  
| Dacron/Gore-Tex Angioplasty   | 4 (8%)        |  
| 10 (20%)                     | 2 (6%)        |  
| 2 (6%)                       |  
| 7 (14%)                      | 2 (4%)        |  
| 4 (8%)                       | 3 (6%)        |  
| 2 (4%)                       | 1 (2%)        |  
| 26 (52%)                     | 19 (38%)      |  
| 2 (4%)                       |  
| Table S1 Coarctation patient clinical demographic information |
Figure S1 Location of Aorta measurements. (A) Level of CMR flow measurement: Ascending Aorta above sinotubular junction. (B) Transverse aortic arch between innominate and left common carotid artery. (C) Narrowest distal aortic arch (repair/recoarctation site). (D) Abdominal aorta at diaphragm. Two metrics were derived: (1) Transverse arch index (TI) = B/D and (2) Coarctation index (CI) = C/D.