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A Simplified Continuity Equation Approach to the Quantification of Stenotic Bicuspid Aortic Valves using Velocity-Encoded Cardiovascular Magnetic Resonance

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ABSTRACT

Objective: **We aimed to compare velocity-encoded cine cardiac magnetic resonance (CMR) with an established echocardiographic method for noninvasive measurement of aortic valve area (AVA) using the continuity equation. *Methods and Results:* Twenty consecutive young adults with stenotic bicuspid aortic valves were examined with CMR and transthoracic echocardiography (TTE). CMR AVA was calculated by the continuity equation, dividing stroke volume by the aortic velocity-time integral (VTI_{Aorta}), the stroke volume measured both by ventricular volume analysis and by phase contrast velocity mapping at 4 levels (1 subvalvar and 3 supravalvar). Stroke volumes measured at all levels correlated well with those from volumetric analysis. The CMR AVAs calculated using volumetric analysis and VTI_{Aorta} from jet velocity mapping correlated and agreed well with TTE AVA measurements ($R^2 = 0.83$). When CMR AVA was calculated more rapidly using volume flow and VTI_{Aorta} both measured from the same trans-jet velocity acquisition, R^2 was 0.74, with a bias and limits of agreement of 0.02 (-0.44, 0.47) cm^2. *Conclusions*: Continuity equation calculation of the AVA using CMR velocity mapping, with or without ventricular volumetric measurement, correlated and agreed well with the comparable and widely accepted TTE approach.**

INTRODUCTION

Timing of intervention in adult patients with aortic valve stenosis is largely based on the severity of stenosis and the presence of symptoms (1). Currently, transthoracic echocardiography (TTE) is the clinical standard for the evaluation of aortic stenosis. Standard TTE parameters of aortic stenosis severity include measurement of peak aortic velocity (V_{max}), mean transaortic pressure gradient, and continuity equation aortic valve area (AVA) (2). Continuity equation AVA (actually area of the vena contracta) is calculated from the principle that volume flow proximal to the valve equals volume flow through the narrowed orifice. Measurements of AVA are achieved by recording two velocity-time integrals (VTIs) from Doppler velocity waveforms acquired proximal and distal to the valve, along with a measurement of the cross sectional area of the left ventricular outflow tract (LVOT) (3). Several studies have shown good concordance of echocardiographic and invasive catheter-based estimates of aortic valve area (3–5).

To date, most cardiac magnetic resonance (CMR) studies for the quantification of stenotic aortic valves have focused on direct planimetry of the stenotic aortic valve (6–12), although the dimensions of CMR voxels relative to the size and shape of the stenotic orifices make this approach questionable, especially in severe or irregularly shaped stenoses. Only one previous study has investigated the use of velocity-encoded cine

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magnetic resonance (VEC-MR) for calculation of AVA by the continuity equation (13). In the standard continuity equation, the stroke volume is estimated by multiplying VTI_{LVOT} by the cross-sectional LVOT area. But this is not the only possible approach by CMR since phase contrast velocity mapping can also determine the stroke volume more directly by measuring the volume of flow through planes transecting the LVOT, the stenotic jet or the proximal ascending aorta. Furthermore, the stroke volume calculated from standard left ventricular analysis can be used, as long as there is no significant mitral regurgitation. In theory, it is thus possible to determine AVA using the continuity equation [AVA = stroke volume/VTI_{Aorta}] with direct input of stroke volumes instead of LVOT areas and VTI_{LVOT}.

The purpose of the present study was to compare CMR measurements of AVA using the continuity equation with those calculated by echocardiography, the LV stroke volumes being measured by CMR either by volumetric analysis or by direct flow measurement at one or more levels.

METHODS

Patient population

We prospectively recruited 20 consecutive young adult patients (13 men, age 35 ± 4 years) with aortic stenosis (all bicuspid aortic valves), who visited the outpatient clinic for Adult Congenital Heart Disease of the Thoraxcenter from December 1, 2005 until September 25, 2006. The only exclusion criteria were those for general CMR suitability (14) and the presence of a significant mitral regurgitation. The institutional review board approved the study protocol, and all patients gave written informed consent.

Transthoracic echocardiography

Patients were imaged with TTE by a single experienced sonographer (JMG), with the use of a commercially ultrasound system (Sonos 7500, Philips Medical Systems, Best, The Netherlands) with a 2 to 4-MHz, 128-element, phased-array transducer. Images were acquired by using standard imaging windows with short breath-holds used as needed. Doppler flow data were acquired from the LVOT region in pulsed wave mode and from the aortic valve in continuous wave mode. The LVOT diameter was measured in the parasternal long-axis view in midsystole at the same level as the LVOT pulsed wave Doppler velocity measurement, and then converted to LVOT area [LVOT area = π *(0.5*LVOT diameter)^2].

Peak velocities and VTIs were used to calculate AVA according to the continuity equation [AVA = stroke volume/VTI_{Aorta}], where the Doppler-derived stroke volume is calculated by [SV = LVOT area* VTI_{LVOT}]. Calculations were based on the single best representative heartbeat as selected independently by the expert sonographer (JMG), blinded to the CMR data.

Cardiovascular magnetic resonance imaging

A clinical 1.5-Tesla MRI scanner with a dedicated cardiac eight-element phased-array receiver coil was used for imaging (Signa CV/I, GE Medical Systems, Milwaukee, Wisconsin, USA). Electrocardiographic gating was used, with cine images acquired during expiratory breath-holds. A standard ventricular function examination was performed by initial acquisition of steady-state free-precession (SSFP) cine images in standard long axis planes (2- and 4-chamber view, and LVOT view) by one experienced CMR cardiologist (RJMvG). The following imaging parameters were used: 6–10 s per breath-hold per slice (depending on heart rate); 24 phases per slice location; field of view (FOV), 300 × 340 mm^2; repetition time, 3.0 to 3.4 ms; echo time, 1.5 ms; flip angle (α), 45°; matrix, 224 × 256. To cover the entire left ventricle, 9–12 short axis slices, 8 mm slice thickness with 2 mm gap, were acquired perpendicular to the 4-chamber long-axis view of the left ventricle.

For the quantitative flow measurements, a retrospectively gated phase contrast sequence was used during expiratory breath-holds (<15 seconds), velocity encoded through-plane in the slice select gradient direction. The following imaging parameters were used: 30 frames/heartbeat; FOV, 400 × 200 mm^2; repetition time, 6.5 ms; echo time, 3.1 ms; flip angle (α), 30°; matrix, 256 × 128. The LVOT cine views were used to locate four velocity-mapping planes parallel to the aortic valve plane at its greatest excursion toward the apex (typically, end-systole). One plane was located at the tips of the aortic valve, and the other 3 were positioned parallel to this, but offset +12, +6, and −18 mm (Fig. 1). The naming convention adopted for these slices is as follows, moving from the aorta towards the LVOT: Level ++, Level +, Level 0, and Level -. Typically, the velocity encoding range (V_{ENC}) was 2.0 m/s for the LVOT (Level −) and 4.0 m/s in the aorta (Levels 0, + and ++). For the aorta, V_{ENC} was initially 4.0 m/s but was increased a priori if the peak velocity could be predicted to be greater, based on the patient's functional images, for example, when they showed limited valve opening or a narrow systolic jet. Flow images were re-acquired with a higher V_{ENC} if velocity aliasing occurred.

For the quantitative flow measurements, the data were transmitted to an offline image processing station (Cine version 3.4, GE Medical Systems, Milwaukee, Wisconsin, USA), and the quantitative flow images were analyzed by an independent observer (SCY) unaware of the echocardiographic data.

For each of the 4 CMR imaging levels, regions of interest (ROIs) were drawn on each of the 30 frames of the cine to include the lumen of the LVOT, the aortic valve, or the aorta, depending on slice position. Peak velocities were determined by extracting the greatest velocity recorded in any pixel within the ROI across the valve flow field. Flow and peak velocity data (Fig. 2) were exported to a spreadsheet and stroke volumes and VTIs determined. VTIs and stroke volumes were calculated by using Simpson's rule to integrate the peak velocity (cm/s) and aortic flow (mL/s), respectively, versus time (ms) during systole. Only the VTI from the level with the highest V_{max}

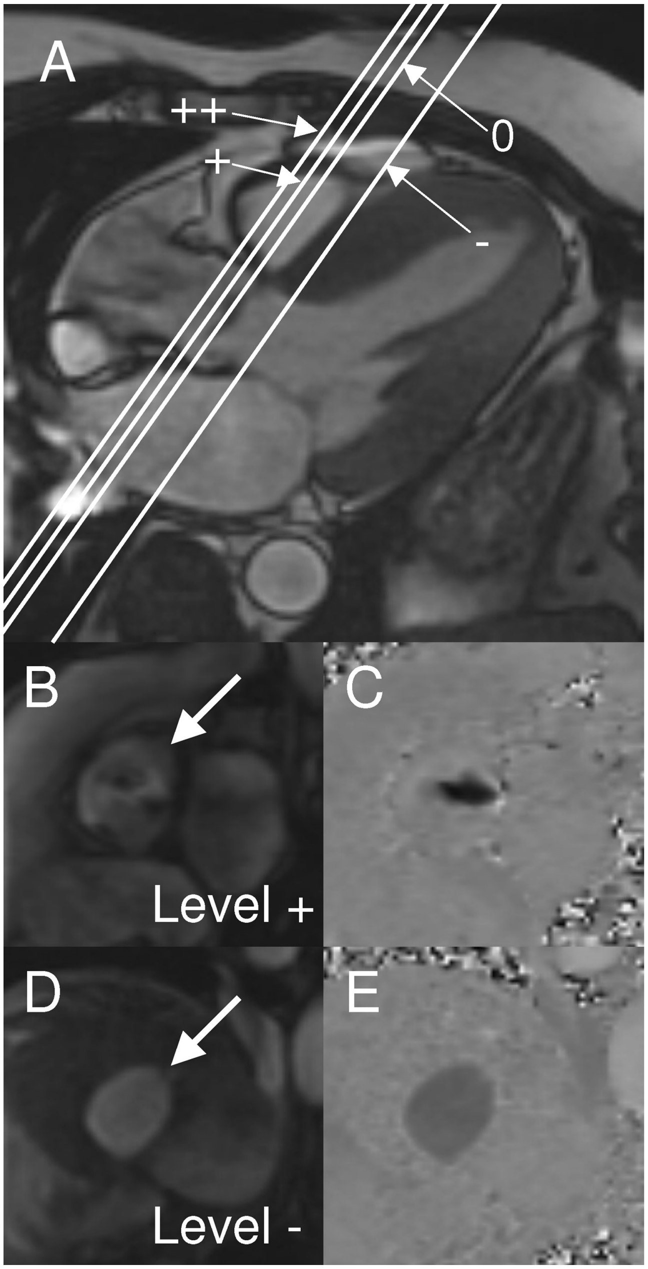

Figure 1. (A) Cine CMR image in the LVOT plane showing the location of the four velocity mapping planes at Levels 0, +, ++ and −. From these, both magnitude (B and D) and corresponding phase images (C and E) were reconstructed. Note the stenotic bicuspid aortic valve (arrow in B) and the oval LVOT tract (arrow in D).

(VTI_{Aorta}) was used for further calculation of AVA and statistical analysis.

CMR AVA was calculated using the continuity equation [$AVA = SV/VTI_{Aorta}$], the stroke volumes being determined by 3 different methods (Table 1). To investigate if stroke volumes measured from the flow images were accurate (method B), these stroke volumes were compared with those calculated by the standard left ventricular analysis (method A) using commercially available software (CAAS-MRV, Pie Medical Imaging, Maastricht, The Netherlands) by one independent observer (RJvG) unaware of stroke volumes measured from the flow images.

Table 1. Three different methods of AVA calculation by CMR.

Method	Formula AVA calculation (SV/VTI_{Aorta})	Description SV determination
A	($LVEDV - LVESV$) / VTI_{Aorta}	SV by LV volumes
B	Flow-time integral aorta / VTI_{Aorta}	SV by VEC-MR (flow data) at aortic level
C	(LVOT area * VTI_{LVOT}) / VTI_{Aorta}	SV by VEC-MR (peak velocity data) at LVOT level

AVA = aortic valve area, LV = left ventricle, LVEDV = left ventricular end-diastolic volume, LVESV = left ventricular end-systolic volume, LVOT = left ventricular outflow tract, SV = stroke volume, VEC-MR = velocity-encoded cine magnetic resonance, VTI = velocity-time integral.

When using method C (comparable to the method used in ultrasound), there are several approaches for the measurement of the LVOT area by CMR. In ultrasound the practiced norm is to measure the diameter of the LVOT and assume the cross-section to be circular. The LVOT diameter was therefore measured from the LVOT view at mid systole at Level− and assumed to be circular for consistency with ultrasound.

According to the American College of Cardiology/American Heart Association (ACC/AHA) guidelines (1), aortic stenosis with a valve area >1.5 cm^2 was considered as mild, 1.0 to 1.5 cm^2 as moderate, and <1.0 cm^2 as severe.

Statistical analysis

All continuous data are expressed as mean ± SD. The correlation between CMR and TTE methods was determined by linear regression analysis, including standard errors of the estimate (SEE). Agreement between techniques was evaluated by the standard paired t-test. Furthermore, Bland-Altman analysis was used to determine the mean of the difference with 95% limits of agreement (±1.96 standard deviation) (15) A two-tailed probability value <0.05 was considered statistically significant. All statistical analyses were performed using SPSS software (version 12.0, SPSS Inc., Chicago, Illinois, USA).

RESULTS

All patients (n = 20) completed the imaging protocols without difficulty. The CMR study was performed 24 ± 14 days after the echocardiographic data acquisition. There was no change in clinical status or medication use between the echocardiographic exam and the CMR study. The patient population showed a wide range in severity of aortic stenosis, with valve areas measured by TTE from 0.80 to 2.28 cm^2 (mean 1.34 ± 0.45 cm^2). Based on TTE, patients were classified as having mild (n = 6), moderate (n = 9), or severe aortic stenosis (n = 5). All patients were in sinus rhythm, and 12 patients (60%) had at least mild

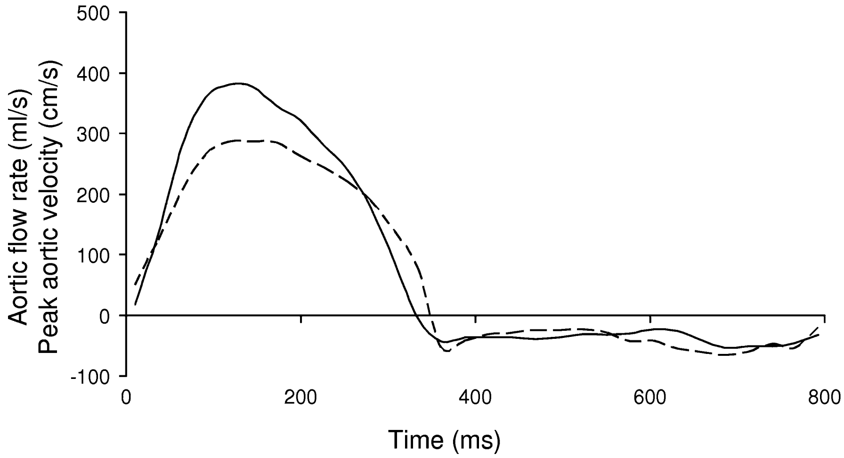

Figure 2. Aortic flow rate (solid line) and peak aortic velocities (dashed line) measured at level +. The area under the curve during systole was used for calculation of stroke volume and VTI_{Aorta}, respectively. Note the presence of aortic regurgitation characterized by a negative flow.

aortic regurgitation (maximal 2+, qualitative assessment). The left ventricular ejection fraction measured by CMR averaged 59 ± 9%, ranging from 42% to 74%.

Flow measurements

Measurements by CMR of peak velocities in the aorta correlated well with those by TTE ($R^2 = 0.88$, SEE 0.30 m/s). The correlation was less good between CMR and TTE measurement at LVOT level ($R^2 = 0.60$, SEE 0.15 m/s). Bland-Altman analysis gave the following mean differences and limits of agreement between CMR and TTE peak velocities: –0.11 (−0.68 to 0.46) m/s at aortic level and 0.12 (−0.17 to 0.41) m/s at LVOT level. The highest peak aortic velocity was found at level 0 (at the tips of the valve) in 9 patients (45%), and at level + in the remainder (55%).

VTIs incorporate more data than peak velocities and are thus more robust. CMR data correlated well with TTE data for VTI_{Aorta} and correlated moderately with TTE for VTI_{LVOT} (Fig. 3). Bland-Altman analysis demonstrated a small but significant difference between CMR and TTE measurements for VTI_{Aorta} and VTI_{LVOT} with relatively large limits of agreement for VTI_{LVOT}.

Stroke volumes measured directly from flow acquisitions at all four levels showed good correspondence with stroke volume measurements by volumetric analysis, with no statistically significant differences (Table 2). The best correlation with volumetric measurements ($R^2 = 0.73$, SEE 12.3 mL) was found when flow volume was measured at level +, with a bias of 0.3 mL and limits of agreement of –23.3 to 23.8 mL. Doppler-derived stroke volumes showed only moderate correlation ($R^2 = 0.42$, SEE 17.2 mL) with CMR volumetric data, with a nonsignificant bias of 6.2 ml and limits of agreement of −28.1 to 40.4 mL.

Table 2. Comparison of stroke volume measurements by VEC-MR at each of the four levels to CMR ventricular volumetric analysis

Stroke volume (mL)	Linear regression analysis R^2	SEE	Slope	Intercept	Bland-Altman analysis Bias	Limits of agreement
Level −	0.64	11.03	0.75	24.6	0.1	−23.0, 23.2
Level 0	0.65	12.91	0.89	11.1	0.8	−24.2, 25.7
Level +	0.73	12.31	1.03	−2.7	0.3	−23.3, 23.8
Level ++	0.64	16.35	1.10	−8.4	1.5	−29.9, 32.9

AVA measurements

Table 3 shows linear regression and Bland-Altman comparisons of the three types of CMR measurement of AVA with those by TTE.

Figure 4A shows the AVAs measured by CMR, with stroke volume and jet velocity measurements derived from the same velocity-encoded acquisition (method B), plotted against the TTE AVA measurements for all 20 patients. The Bland-Altman plot shows a mean bias, CMR-TTE, of only 0.02 cm^2 with relatively small limits of agreements (Fig. 4B).

Measurements of AVA by the standard continuity equation using CMR (method C), assuming a circular LVOT, were larger than by TTE, despite a good correlation (bias = 0.37 cm^2, p = 0.04, $R^2 = 0.69$, SEE 0.36 cm^2).

DISCUSSION

In routine clinical practice, TTE has become the accepted standard for evaluation of aortic valve stenosis, as it is

Table 3. Comparison of AVA calculation by CMR to TTE.

	Linear regression analysis R^2	SEE	Slope	Intercept	Bland-Altman analysis Bias	Limits of agreement
Method A	0.83	0.17	0.81	0.25	−0.01	−0.38, 0.36
Method B	0.74	0.23	0.83	0.25	0.02	−0.44, 0.47
Method C	0.69	0.36	1.19	0.11	0.37*	−0.34, 1.08

*p < 0.001 by paired Student's t test.

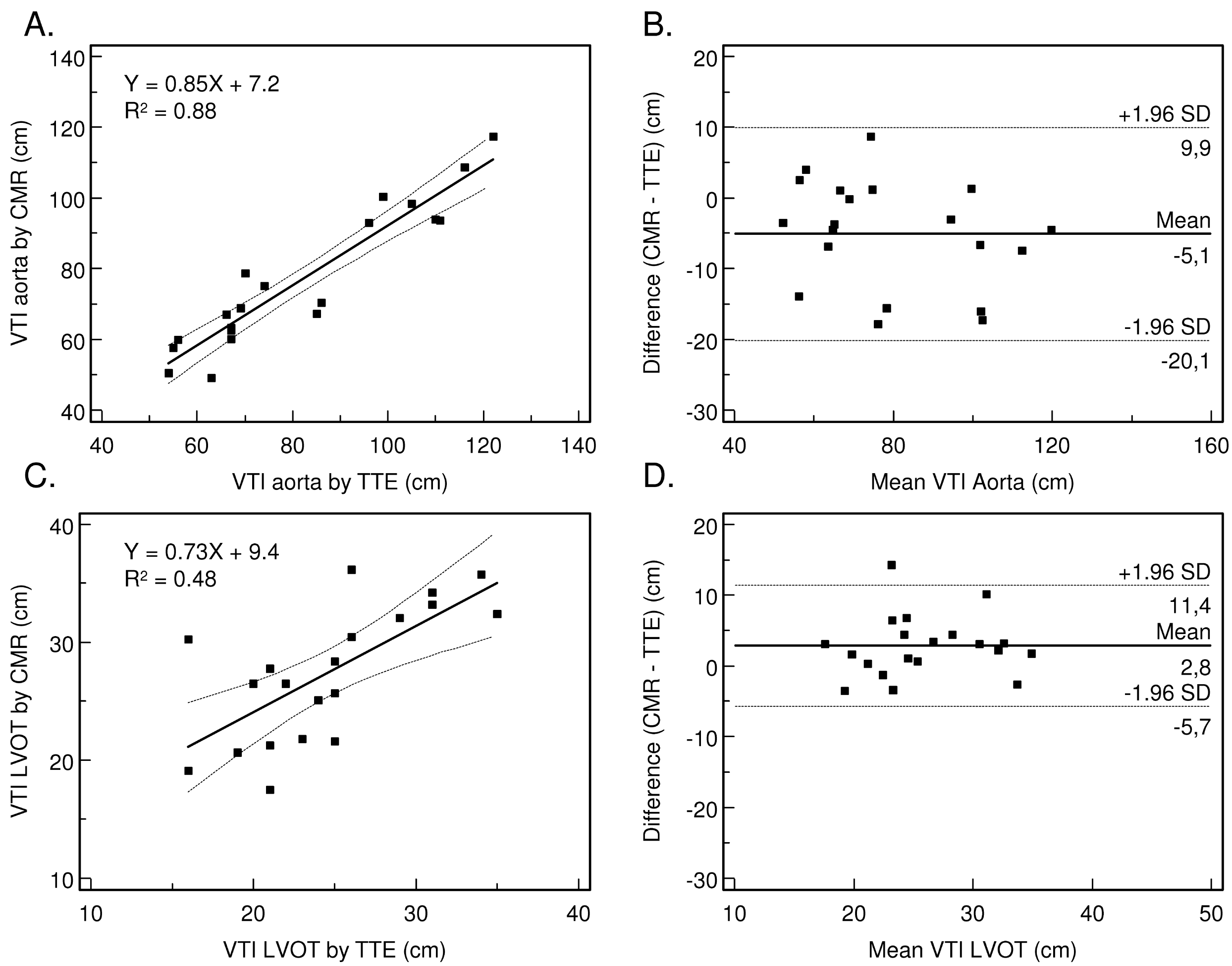

Figure 3. Comparisons between CMR and TTE measurements of peak velocity-time integrals (VTI) at aortic (A and B) and LVOT (C and D) levels. The VTI_{Aorta} measured by CMR correlated well with TTE (A), with slight underestimation as shown by the Bland-Altman analysis (B), while correlation was poorer for VTI_{LVOT} (C), with a slight overestimation by CMR relative to TTE (D).

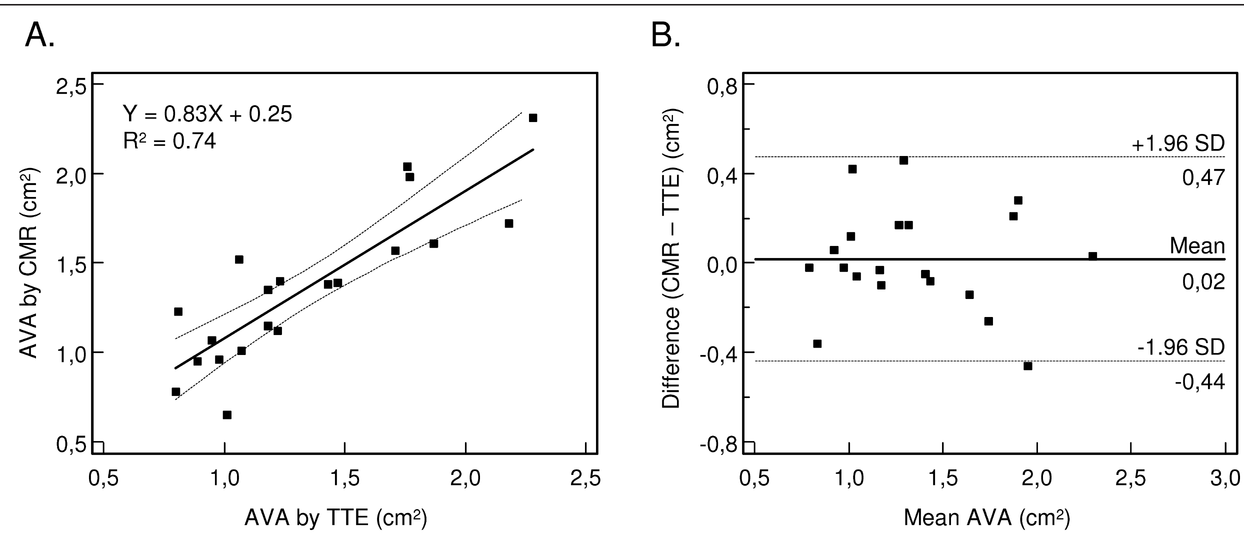

Figure 4. Comparisons between CMR (method B) and TTE measurements of aortic valve areas (AVA). For the CMR measurements the stroke volumes and VTIs were measured from the same CMR velocity map at the level (0 or +), which was found to yield the highest peak velocity.

non-invasive, readily available, fast, and can be performed at reasonable cost. The echocardiographic assessment includes measurement of peak aortic velocity and calculation of continuity equation AVA. Peak velocities, however, are dependent on the volume flow rate, overestimating the severity of stenosis when flow is elevated, as in the presence of aortic regurgitation, or underestimating the severity of stenosis if flow is reduced, i.e., when there is left ventricular systolic dysfunction or mitral regurgitation. Calculation of AVA by the continuity equation has the advantage of taking the flow rate into account. Echocardiographic measurements have prognostic value as most studies of outcomes have used echocardiographic indices (16–18). In most patients the echocardiographic study is adequate for the assessment of stenosis, although in certain situations, TTE may not be reliable due to poor acoustic windows, calcification of the aortic valve limiting accurate measurements of LVOT diameter, eccentric jets (especially in patients with congenital aortic stenosis), and inaccurate jet localization. Where echocardiographic results are inconclusive, a comparable investigation by CMR for congenital or acquired valvular heart disease would be valuable. There may also be cases where CMR is chosen anyway for the investigation of additional congenital or acquired cardiovascular pathology.

Most CMR studies for the quantification of stenotic aortic valves have focused on direct planimetry of the stenotic aortic valve or jet area, and they have shown varied although generally acceptable results (6–12). However, our clinical experience regarding the accuracy of CMR planimetry is not consistent, there being cases where the borders of the orifice, or rather the jet, remain unclear. This is not surprising given the typical CMR slice thicknesses of 5 mm or more, relative large pixel dimensions compared to jet area, and the splayed or fragmented nature of some post stenotic jets. There is also potential for error due to misplacement of the slice due to motion of the valve or inconsistency in the position of breath holds. It is also unclear how signal loss due to parajet shear and turbulence effects edge discrimination.

The use of velocity encoding to determine the peak VTI within a jet may be less subject to these limitations because only the voxel or voxels located fully within the core of the jet are required for determination of the peak velocity and peak VTI, and the jet core will usually extend 10 mm or more beyond the orifice rendering exact plane position less crucial. Furthermore, voxels that span the boundaries of the jet or vessel lumen do contribute to volume flow measurements, but, theoretically at least, partial volume averaging in these voxels should matter less when flow (spatial mean of velocity $\times$ area for each phase of the cycle) rather than jet area is the subject of measurement. This appears to have been the case as the cardiac outputs derived from our CMR flow measurements correlated and agreed well with those by CMR ventricular volumetric analysis in our study population, where no patient had significant mitral regurgitation. Furthermore, although much reluctance exists to measure stroke volume at the level of the highest recorded velocity, the correlation and agreement were as good if not better when stroke volume was measured at this level, presumably at or close to the vena contracta. This may be explained by the relatively laminar flow within the jet at this level, rather than the convergent flow upstream or the divergent and turbulent flow downstream. However, it is important to realize that our patient cohort did not include patients with peak velocities $>$5.0 m/s. At high values phase contrast CMR underestimate velocity (and therefore flow) (13), which is suggested to be due to intra-voxel dephasing characterized by signal loss in the magnitude image. Echo time minimization is essential to reduce error. Other problems associated with CMR velocity data are ghosting artifacts and movement of the aortic valve during contraction.

In our comparative study, CMR results correlated and agreed well with TTE with respect to aortic jet velocity data (i.e. peak velocities and VTIs) and continuity equation AVAs based on volumetric (method A) or aortic flow measurement (method B) of stroke volumes. Previously, Caruthers et al. (13) have shown that CMR correlated well with TTE when the standard continuity equation (method C) was used ($R^2 = 0.69$), by means of the identified best approach, which was confirmed by our study. The present study is, to our knowledge, the first to demonstrate the usefulness and relative advantage of aortic flow measurement of stroke volumes (method B) in determining the continuity equation AVA by CMR. This method is easier to use than the method previously proposed by Caruthers et al., and both are equally accurate (13). The continuity equation AVA using volumetric analysis has been previously proposed for ultrasound (not for CMR) by Dumont et al. (19), demonstrating an additive value of the method, especially in patients with LVOT flow acceleration. Furthermore, Haghi et al. (20, 21) have shown that a hybrid approach, combining stroke volume measured by volumetric CMR data and continuous-wave Doppler data by TTE, for determining continuity equation AVA correlates well with TTE. However, both studies determined stroke volume by the difference between end-diastolic and end-systolic volumes. This method could be incorrect when associated mitral regurgitation is present, whereas the measurements of aortic flow avoid this limitation. Another advantage of aortic flow measurement is that it is easier and faster to analyze than ventricular volumes measured from multiple short axis cine acquisitions. Our findings showed that estimates of stroke volumes were accurate at all 4 levels, and that the highest peak aortic velocities were acquired at 2 levels (level 0 or +). Therefore, using the modified continuity equation, only 2 flow velocity maps will be necessary to estimate AVA.

There are several reasons why the simplified continuity equation (method A or B) is theoretically preferable to the standard continuity equation (method C) when using CMR. First, reliable measurements of LVOT peak velocity and area are needed for method C. This represents a potential source of error, in part because of the converging boundaries and ovoid cross section of the outflow tract. Waters et al. (22) have shown that VTI measurements at different levels in the post stenotic jet were comparable, indicating relative insensitivity to the position of the imaging plane in the aorta. In the LVOT, however, measurements were position-dependent, observing higher values when the image plane was close to the valve, which would explain the lower correlation between CMR and ultrasound measurements of LVOT velocities and VTIs in our study. Secondly, the assumption that the LVOT area is circular is incorrect (see Fig. 1D). Baumgartner et al. (23) have shown that by using the standard continuity equation, and, assuming a circular LVOT, AVA is significantly underestimated by TTE compared to cardiac catheterization using the Gorlin formula. By using direct planimetry of the LVOT by TTE, results were more consistent with invasive data. By using method A or B, no assumptions regarding stroke volumes are required and the problems associated with determination of LVOT velocity and area are avoided. In spite of the limitations of the standard continuity equation used by TTE, however, we found relatively good correlation and agreement between AVAs determined by this approach and by CMR using the simplified continuity equation. No clear explanation exists for this finding.

Study limitations

First, our study lacks a reliable gold standard because the accuracy of the results obtained by TTE remains uncertain. More data on the correlation between the described methods and invasive catheterization data (using the Gorlin formula) and other imaging modalities (for example direct planimetry by CT or transesophageal echocardiography) would be valuable.

Second, we studied a relatively selected patient population of young adults with congenital aortic stenosis. However, the approach used should also be applicable to older patients with degenerative aortic stenosis in whom calcification is unlikely to compromise stroke volume and VTI measurements as much as it does planimetric approaches. Third, our study protocol did not include repeated measurements of one patient; therefore, no reproducibility data are available. Finally, due to the small sample size, all conclusions of the present study must be drawn with caution.

CONCLUSIONS

We conclude that the use of the continuity equation for determining AVA by velocity-encoded CMR is feasible, simple and compares well with the established echocardiographic approach in patients with stenotic bicuspid aortic valves, providing an attractive alternative non-invasive approach to the quantification of aortic stenosis severity.

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