where $\gamma$ is the gyromagnetic ratio, and $\Delta m$ corresponds to the difference of the first moment of the gradient–time curve. The closer the VENC is to the maximum expected velocity (ideal VENC), the more precise is the measurement. The pulse sequences are capable of encoding velocity perpendicular (through-plane measurements) or parallel (in-plane measurements) to the direction of flow (Fig. 2). Commercial software programs are widely available for quantitative analysis of through-plane PC-MRI. In-plane measurements, in which flow encoding can be performed in almost any direction parallel to the imaging plane, often are useful for localizing stenotic lesions and, if available, can be quantified in the same manner as through-plane measurements.

There are several potential sources of error with PC-MRI measurements. A VENC setting below the peak velocity results in aliasing, or velocity wraparound. As the phase shift of this peak velocity exceeds $180^\circ$, the signal intensity adjacent to areas of maximum brightness decreases and vice versa if peak velocity exceeds $-180^\circ$ (Fig. 1C). However, a VENC that is too high results in less accurate measurements because noise increases with larger VENCs, and flow becomes compressed to a narrow range (Fig. 1D). Large deviations of VENC from ideal VENC affect peak velocity more than flow because the noise is averaged over a number of voxels for flow measurements. Peak velocity measurements show deviation less than 10%, a clinically acceptable level of error, if the VENC is set at no more than three times

**OBJECTIVE.** The purpose of this study was to review and illustrate various clinical applications of phase-contrast MRI.

**CONCLUSION.** Cardiac MRI has emerged as a valuable noninvasive clinical tool for evaluation of the cardiovascular system. Phase-contrast MRI has a variety of established applications in quantifying blood flow and velocity and several emerging applications, such as evaluation of diastolic function and myocardial dyssynchrony.
the ideal VENC [1]. In general, normal arterial flow velocities are less than 150 cm/s, normal pulmonary venous flow velocities are less than 100 cm/s [3], and normal systemic venous flow velocities are less than 50 cm/s [4, 5].

In addition to improper VENC setting, sources of error in PC-MRI measurement include deviation of the imaging plane during data acquisition (e.g., cardiac or respiratory motion), inadequate temporal resolution, inadequate spatial resolution, and field inhomogeneity (e.g., susceptibility artifact from metallic implants). In addition, the use of retrospective gating, as opposed to prospective (triggered) gating, is important for evaluation of diastolic events (e.g., diastolic function, semilunar valve regurgitation) to ensure capture of complete diastole. Depending on the structure of interest, PC-MRI parameters should be set to minimize potential sources of error.

Velocity and flow are measured with commercial software that allows users to define regions of interest around a vessel lumen or intracardiac region sampled throughout the cardiac cycle. In comparison with Doppler sonography, PC-MRI flow measurements tend to show lower peak velocities [6], but studies have shown that at Doppler sonography, peak velocities can be overestimated as much as 25% [7]. Mean flow in large vessels also can be overestimated owing to an assumption of constant velocity over the whole vessel area. In contrast, at PC-MRI, one can take into account the variation of flow within the vessel. PC-MRI therefore has been found superior to Doppler sonography for evaluation of mean flow [8].

Clinical Applications

Evaluation of Vascular Flow

Aortic coarctation—Aortic coarctation, considerable narrowing of the aorta, commonly near the left subclavian artery origin, can be evaluated with PC-MRI in several ways. First, flow turbulence can be assessed qualitatively with in-plane PC-MRI aligned along the flow jet (Fig. 2). Severity can be evaluated by measurement of the pressure decrease \( \left( P_{\text{distal}} - P_{\text{proximal}} \right) \) across the coarctation site, which can be estimated with the modified Bernoulli equation, or

\[
\left( P_{\text{distal}} - P_{\text{proximal}} \right) = 4 \times V_{\text{max distal}}^2,
\]

where \( V_{\text{max distal}} \) is the peak stenotic velocity (m/s) immediately distal to the narrowing, \( V_{\text{max proximal}} \) is the peak velocity (m/s) immediately proximal to the narrowing, and \( K \) is the loss, or Bernoulli coefficient (mm Hg·m⁻²), commonly taken as 4.0. The modified Bernoulli equation can be simplified to

\[
(P_{\text{distal}} - P_{\text{proximal}}) = 4 \times V_{\text{max distal}}^2
\]

when the proximal velocity is small (< 1.0 m/s for arterial flow) compared with the distal velocity.

Derivation of the modified Bernoulli equation is based on several assumptions, including the presence of nonviscous, steady, incompressible flow. Inaccuracies can lead to overestimation of the pressure difference in mild stenosis (≤ 60%) because proximal velocity is ignored and distal pressure recovery is assumed to be small, and to underestimation of the pressure difference in severe or long stenosis [9] because of increased turbulence downstream with increased pressure losses and the presence of increased viscous forces. Inclusion of proximal velocity and adjustment of the loss coefficient \( K \) for the anatomic degree of stenosis improve estimation of the pressure difference across the area of stenosis [10]. The presence and degree of collateral flow can be assessed by comparison of aortic flow immediately distal to the coarctation and more distally at the level of the diaphragm [11]. A resting pressure difference greater than 30 mm Hg across the coarctation and the presence of collateral vessels, which can lower the resting gradient, are indications for intervention [12].

Aortic dissection—PC-MRI can be helpful for the diagnosis of aortic dissection [13] and for elucidation of true and false lumens and entry points into a false lumen. This technique can be particularly useful to patients in whom the use of gadolinium chelates is relatively contraindicated, such as patients with renal insufficiency who are at risk of nephrogenic systemic fibrosis [14]. In general, peak and mean velocities are lower in the false lumen, and bidirectional flow with greater retrograde flow volume is more often found in the false than in the true lumen [15] (Fig. 3).

Pulmonary artery—Patients with congenital heart disease often have involvement of the pulmonary arteries, such as stenosis and hypoplasia, or problems with surgical conduits to the pulmonary arteries with resultant hyperperfusion of the affected lungs. With PC-MRI, differential pulmonary blood flow is quantified by measurement of flow separately in the left and right pulmonary arteries [16] (Fig. 4) or by measurement of peak pressure differences across areas of stenosis by performance of PC-MRI and application of the modified Bernoulli equation.

Cerebrovascular arteries—PC-MRI evaluation of the cerebrovascular system can yield adjunctive information to routine MR angiography on the hemodynamic effect of carotid atherosclerotic disease [17] and the presence of collateral flow, vertebrobasilar insufficiency, and arteriovenous malformations [18, 19] (Fig. 5).

Quantification of Cardiac Function

Cardiac output—PC-MRI can be performed to calculate cardiac output [20, 21] and in the presence of clinically significant valvular regurgitation enables more accurate assessment of cardiac output than does cine MRI. With an imaging plane through the aortic root, net forward volume, volume during systole, and stroke volume can be accurately measured. Multiplying stroke volume by heart rate gives cardiac output. Similarly, use of a cross-sectional imaging plane through the main pulmonary artery results in accurate measurement of the cardiac output of the right ventricle.

Diastolic function—Impaired left ventricular filling, diastolic dysfunction, has been found to correlate with poor prognosis [22]. Diastolic function can be measured with retrospective cardiac-gated PC-MRI in a variety of ways similar to the methods used in Doppler echocardiography [23, 24]. Measurement of flow across the mitral valve yields information on early (E) and late or atrial (A) left ventricular filling patterns (Figs. 6–9). The addition of pulmonary vein systolic (X), early diastolic (Y), and atrial systolic (Z) peak flow velocities (Figs. 6–9) and longitudinal myocardial early (E') and late (A') diastolic flow velocities obtained at the base of the heart can improve evaluation of diastolic function. A ratio of \( E \) to \( E' > 15 \) is considered diagnostic of the presence of diastolic dysfunction and has been correlated with elevated left ventricular filling pressures and poor prognosis [22].

Cardiac dyssynchrony—Cardiac resynchronization therapy has proved beneficial to patients with heart failure who have evidence of left ventricular dyssynchrony. PC-MRI with techniques for measuring the lower velocities of myocardial wall motion can be used to obtain multidirectional velocity information similar to that obtained with tissue Doppler imaging without the limitations imposed by acoustic windows [25, 26] (Fig. 10).
Evaluation of Congenital Shunt Lesions

PC-MRI is a valuable tool for evaluating flow dynamics in congenital heart disease. Directionality and quantification of flow in anomalous vascular structures can be directly measured (Fig. 11). Intracardiac shunts can be quantified by measurement of flow in the pulmonary trunk and aorta and calculation of the pulmonary to systemic blood flow ratio (Qp/Qs). In the absence of a shunt, pulmonary and systemic flow rates are approximately equal, with a ratio of 1.0. In left-to-right shunts, as in atrial septal defect, flow in the pulmonary artery is greater than flow in the aorta. In patent ductus arteriosus, ascending aortic blood flow is greater than main pulmonary arterial flow, and shunt volume can be calculated by subtraction of the two flows.

In addition to measurement of Qp/Qs, the direct contribution of anomalous veins, and atrial or ventricular septal defects can be evaluated individually with through-plane PC-MRI. In-plane PC-MRI at the level of a defect can show the presence and direction of flow through the defect more clearly than routine cine MRI and may show evidence of bidirectional flow (Fig. 12).

Evaluation of Valvular Disease

The severity of valvular disease can be assessed and quantified with PC-MRI (Fig. 13); the method has been validated in several studies [27–29]. Flow measurements with PC-MRI show good correlation with Doppler echocardiographic findings, although peak velocity measurements are commonly underestimated with PC-MRI [30, 31]. For regurgitation, forward and reverse flow measured during a cardiac cycle can be used to calculate the regurgitant fraction, defined as follows: (reverse flow volume / forward flow volume) × 100%. For aortic and pulmonary valve regurgitation, flows are measured in the proximal aorta and main pulmonary artery, respectively.

However, for mitral and tricuspid valve regurgitation, comparison of flow measurements obtained between systole and diastole is difficult because of considerable annulus motion (up to 2 cm) [32, 33] during the cardiac cycle, which necessitates correction for through-plane motion [33–35]. The regurgitant fraction can instead be ascertained with the following formula: [(stroke volume – forward flow) / stroke volume] × 100% [36]. Forward flow is obtained with PC-MRI of the great vessel, and stroke volume is calculated with end-diastolic and end-systolic ventricular volumes from cine imaging.

Severity of stenosis on PC-MRI images can be calculated by measurement of the pressure difference across the stenotic valve with the modified Bernoulli equation. However, the orientation of the imaging plane should be perpendicular to the flow jet rather than the vessel to minimize error in peak velocity measurement caused by oblique positioning [29].

Other

PC-MRI can be used for evaluation of the midcavity and outflow tract flow disturbances that commonly occur in hypertrophic hearts and subaortic ring. In-plane PC-MRI views show flow disturbances (Fig. 14), and through-plane PC-MRI at specific levels can be used to quantify the degree of stenosis at each point. In this manner, the degree of stenosis contributed by midcavity versus outflow tract versus valvular lesions can be differentiated.

Acknowledgment

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Fig. 1—15-year-old boy with aortic coarctation.
Through-plane phase-contrast MR images (TR, 39.6; acquisition time, 23 seconds; voxel size, 2.6 × 1.3 × 6 mm; retrospective gating) with three velocity-encoding (VENC) value settings at level just distal to coarctation site. A, Magnitude image. B, Phase-contrast MR image at ideal VENC (300 cm/s; peak, 220 cm/s) shows flow pattern in descending aorta is well defined with no aliasing. Signal intensity is high in ascending aorta (black arrow) and low in descending aorta (white arrow), indicating opposite directions of flow. C, Phase-contrast MR image at lower VENC (200 cm/s; peak, 200 cm/s) than in B shows aliasing of flow in descending aorta (arrows). Signal intensity is high in ascending aorta (black arrow) and low in descending aorta (white arrow), indicating opposite directions of flow. D, Phase-contrast image at higher VENC (600 cm/s; peak, 201 cm/s) than in B shows flow in descending aorta is not well defined (arrows). Signal intensity is high in ascending aorta (black arrow) and low in descending aorta (white arrow), indicating opposite directions of flow.
Fig. 2—30-year-old man with bicuspid aortic valve, dilated ascending aorta, and mild coarctation of descending aorta.

A, Magnitude image.

B, In-plane phase-contrast image (TR, 51; acquisition time, 23 seconds; voxel size, 1.9 × 1.2 × 5 mm; velocity-encoding value, 400 cm/s; retrospective gating) oriented in oblique sagittal plane shows change in flow pattern at level of coarctation (dashed line) with flow directed anteriorly in proximal descending aorta.

C and D, Through-plane phase-contrast image (C) (TR, 51; acquisition time, 24 seconds; voxel size, 2.0 × 1.3 × 5 mm; velocity-encoding value, 400 cm/s; retrospective gating) shows descending aorta at level just below coarctation (circle) and was used to generate corresponding peak velocity graph (D). In this patient, peak velocity was measured at 270 cm/s with estimated pressure decrease of 29 mm Hg across coarctation, estimated with modified Bernoulli equation. Solid line indicates data; dashed line, spline ± 1.
Fig. 3—60-year-old man with chronic descending aortic dissection. A–D, Magnitude images (A and C), in-plane phase-contrast image (B) (TR, 60; acquisition time, 19.0 seconds; voxel size, 2.8 × 1.4 × 6 mm; velocity-encoding value, 150 cm/s; retrospective gating), and through-plane phase-contrast image (D) (TR, 60; acquisition time, 19.1 seconds; voxel size, 2.8 × 1.4 × 6 mm; velocity-encoding value, 150 cm/s; retrospective gating) show descending aorta. High-velocity flow is evident in smaller of two aortic lumens, which is true lumen (T). F = false lumen.

E, Graph of mean velocities over cardiac cycle in true (T) and false (F) lumens shows high velocity in true lumen and substantial retrograde flow (arrows) only in false lumen. Solid line indicates data; dashed line, spline ± 1.
Fig. 4—2-year-old girl with tetralogy of Fallot.
A, Gradient-echo image at level of pulmonary artery bifurcation shows marked narrowing at origin of left pulmonary artery (arrow). Solid line indicates proximal right pulmonary artery; dashed line, proximal left pulmonary artery.
B, Flow curves from through-plane phase-contrast imaging (TR, 44; acquisition time, 23.8 seconds; voxel size, 2.0 × 1.0 × 6 mm; retrospective gating) performed at level of proximal right pulmonary artery (solid line) (velocity-encoding value, 400 cm/s) and proximal left pulmonary artery (dashed line) (velocity-encoding value, 600 cm/s) show markedly decreased overall flow in left pulmonary artery compared with right pulmonary artery.

Fig. 5—75-year-old man with recent transient ischemic attack and moderate narrowing of proximal left internal carotid artery (LICA).
A, Magnitude image.
B and C, Through-plane phase-contrast image (TR, 49 seconds; acquisition time, 13.6 seconds; voxel size, 2.3 × 1.2 × 6 mm; retrospective gating) (B) and corresponding flow graph (C) show no significant difference in flow between right internal carotid artery (RICA) compared with LICA. Measured peak velocity in RICA was 37.8 cm/s and in LICA was 37.4 cm/s (data not shown). Calculated mean flow in RICA was 312 mL/min and in LICA was 360 mL/min.
Fig. 6—57-year-old man with cardiomyopathy. Imaging planes and regions of interest used in evaluation of cardiac diastolic function. A–F. Gradient-echo (A, C) and spin-echo (E) images show orientation of through-plane phase-contrast imaging (dashed line, A, C, E) for generating through-plane phase-contrast imaging planes with corresponding regions of interest drawn at level of mitral valve leaflet tips (B), basal lateral left ventricular myocardium (D), and left pulmonary vein (F) for evaluation of diastolic function. ECG retrospectively gated acquisition was used to include all diastolic events, and temporal resolution was generally set for 30–50 milliseconds, depending on heart rate and acquisition time. Presence of through-plane annular motion may lead to underestimation of flow at mitral leaflet tips and cause misregistration of lateral wall motion due to deviation of imaging plane.
Fig. 7—25-year-old woman with normal cardiac anatomy and normal systolic and diastolic function and corresponding tissue dynamics measured with phase-contrast MRI and Doppler echocardiography.

A, Time–velocity curve generated from through-plane phase-contrast imaging at level of mitral valve leaflet tips (TR, 42.5; acquisition time, 18.04 seconds; voxel size, 2.8 × 1.4 × 6 mm; velocity-encoding value, 150 cm/s; retrospective gating) shows normal diastolic flow pattern with early (E) velocity (early left ventricular filling) higher than atrial (A) velocity (late left ventricular filling).

B, Transthoracic Doppler echocardiographic image corresponding to A. E = early, A = atrial.

C, Time–velocity curve generated from through-plane phase-contrast imaging at basal myocardial level (TR, 47; acquisition time, 65 seconds; voxel size, 2.8 × 1.4 × 6 mm; velocity-encoding value, 50 cm/s; retrospective gating; number of signals averaged, 3) with measurement of velocities in lateral wall shows normal pattern of early myocardial motion (E' velocity) higher than late myocardial motion (A' velocity).

D, Transthoracic Doppler echocardiographic image corresponding to C. E' = early myocardial, A' = late myocardial.

E, Time–velocity curve generated from through-plane phase-contrast imaging of left inferior pulmonary vein near its ostium (TR, 45; acquisition time, 17.8 seconds; voxel size, 2.8 × 1.4 × 6 mm; velocity-encoding value, 100 cm/s; retrospective gating) shows normal pattern of pulmonary vein flow with higher systolic (X) than diastolic (Y) component and small atrial (Z) reversal.

F, Transthoracic Doppler echocardiographic image corresponding to E. X = systolic, Y = diastolic, Z = atrial.
Fig. 8—85-year-old man with long-standing history of hypertension, mild left ventricular hypertrophy, normal systolic function, and stage I diastolic dysfunction (impaired early left ventricular relaxation).

A. Time–velocity curve generated from through-plane phase-contrast imaging performed at mitral valve leaflet tips (TR, 44; acquisition time, 19.4 seconds; voxel size, 2.7 × 1.5 × 6 mm; velocity-encoding value, 300 cm/s; retrospective gating) shows reduced left ventricular filling in early diastole evidenced by diastolic flow pattern with early (E) velocity (early left ventricular filling) lower than atrial (A) (late left ventricular filling) velocity.

B. Transthoracic Doppler echocardiographic image corresponding to A. E = early, A = atrial.

C. Time–velocity curve generated from through-plane phase-contrast imaging at basal myocardial level (TR, 44; acquisition time, 55 seconds; voxel size, 2.7 × 1.5 × 6 mm; velocity-encoding value, 50 cm/s; retrospective gating; number of signals averaged, 3) with measurement of velocities in lateral wall shows abnormal pattern of early myocardial motion (E’ velocity) < 8 cm/s similar to late myocardial motion (A’ velocity).

D. Transthoracic Doppler echocardiographic image corresponding to C. E’ = early myocardial, A’ = late myocardial.

E. Time–velocity curve generated from through-plane phase-contrast imaging of left inferior pulmonary vein near its ostium (TR, 44; acquisition time, 18.6 seconds; voxel size, 2.7 × 1.5 × 6 mm; velocity-encoding value, 100 cm/s; retrospective gating) shows normal left atrial filling pressures with higher systolic (X) than diastolic (Y) component but prominent atrial (Z) reversal.

F. Transthoracic Doppler echocardiographic image corresponding to E. X = systolic, Y = diastolic, Z = atrial.
Fig. 9—65-year-old man with ischemic cardiomyopathy, severe systolic dysfunction, and stage III diastolic function (restrictive filling).

A, Time–velocity curve generated from through-plane phase-contrast imaging performed at mitral valve leaflet tips (TR, 48; acquisition time, 20.4 seconds; voxel size, 2.7 × 1.4 × 6 mm; velocity-encoding value, 300 cm/s; retrospective gating) shows early (E) to late (A) left ventricular filling velocity ratio > 1.5 with fast deceleration time of early velocity < 160 milliseconds.

B, Transthoracic Doppler echocardiographic image corresponding to A. E = early, A = atrial.

C, Time–velocity curve generated from through-plane phase-contrast imaging at basal myocardial level (TR, 48; acquisition time, 62 seconds; voxel size, 2.7 × 1.4 × 6 mm; velocity-encoding value, 50 cm/s; retrospective gating; number of signals averaged, 3) measurement of velocities in lateral wall shows abnormal pattern of early myocardial motion (E’ velocity) < 8 cm/s is lower than late myocardial motion (A’ velocity) and E/E’ ratio > 15.

D, Transthoracic Doppler echocardiographic image corresponding to C. E’ = early myocardial, A’ = late myocardial.

E, Time–velocity curve generated from through-plane phase-contrast imaging of left inferior pulmonary vein near its ostium (TR, 48; acquisition time, 21.5 seconds; voxel size, 2.7 × 1.4 × 6 mm; velocity-encoding value, 100 cm/s; retrospective gating) shows abnormal left atrial filling pattern with higher diastolic (Y) than systolic (X) component and prominent atrial (Z) reversal.

F, Transthoracic Doppler echocardiographic image corresponding to E. X = systolic, Y = diastolic, Z = atrial.
Fig. 10—64-year-old man with nonischemic cardiomyopathy and marked myocardial dyssynchrony.
A and B, Through-plane phase-contrast image (TR, 34; acquisition time, 62 seconds; voxel size, 2.7 × 1.4 × 6 mm; velocity-encoding value, 50 cm/s; retrospective gating; number of signals averaged, 31) (A) and corresponding myocardial velocity–time curves (B) of septal (S) and lateral (L) walls show systolic (early negative direction velocities) and diastolic (mid and late positive and negative direction velocities) myocardial motion. Marked systolic mechanical dyssynchrony of 90 milliseconds is present between septal wall, which exhibits time to peak systolic velocity at 38 milliseconds, and lateral wall, which exhibits time to peak systolic velocity at 128 milliseconds. Myocardial velocity–time curves also were drawn for anterior (A, A) and inferior (I, A) walls, but time to peak systolic velocity was intermediate between that of septal and lateral walls (not shown).

Fig. 11—6-year-old boy with isolated left subclavian artery.
A, Three-dimensional volume-rendered MR angiographic image shows blind-ending diverticulum (arrow) at presumed origin of left subclavian artery (LSA, arrow), which instead reconstitutes distally.
B, Magnitude image. RIJ = right internal jugular vein, RCC = right common carotid artery, RV = right vertebral artery, LCC = left common carotid artery, LIJ = left internal jugular vein, LV = left vertebral artery.
C, Through-plane phase-contrast (TR, 34; acquisition time, 55 seconds; voxel size, 1.9 × 1.0 × 5 mm; velocity-encoding value, 250 cm/s; retrospective gating) at level of lower neck shows flow in right common carotid (RCC), right vertebral (RV), and left common carotid (LCC) arteries is in same direction, shown as white. However, flow in left vertebral (LV) artery is reversed and in same direction as that in right internal jugular (RIJ) and left internal jugular (LIJ) veins, shown as black.
Fig. 12—25-year-old woman with large secundum atrial septal defect.
A, Magnitude image shows large defect (arrow). B and C, In-plane phase-contrast images (TR, 45; acquisition time, 55 seconds; voxel size, 2.2 × 1.3 × 6 mm; velocity-encoding value, 100 cm/s; retrospective gating) at systole (B) and diastole (C). Large defect is present in atrial septum (arrow) with right-to-left shunting during systole (white, B) and left-to-right shunting during diastole (black, C).

Fig. 13—28-year-old woman with history of balloon angioplasty as child for pulmonic valve stenosis who now has severe pulmonic valve regurgitation.
A–D, Magnitude image (A) and in-plane phase-contrast images (TR, 56; acquisition time, 21.4 seconds; voxel size, 1.9 × 1.2 × 6 mm; velocity-encoding value, 300 cm/s; retrospective gating) in diastolic phase (B), through-plane phase-contrast image (TR, 52; acquisition time, 22.5 seconds; voxel size, 2.1 × 1.3 × 6 mm; velocity-encoding value, 300 cm/s; retrospective gating) at level of main pulmonary artery (dashed line, A and B), and flow graph (D) show dilated pulmonary artery with marked diastolic reverse flow (arrow; B) at level of pulmonary valve. Calculation of flow volume in forward (systole, 115 mL) and reverse (diastole, 52 mL) directions yields regurgitant flow fraction (52 mL / 115 mL) of 45%, indicating severe pulmonic valve regurgitation.
Fig. 14—28-year-old woman with hypertrophic obstructive cardiomyopathy. A–C, Magnitude (A) and in-plane phase-contrast (TR, 55; acquisition time, 24.7 seconds; voxel size, $2.0 \times 1.2 \times 6 \text{ mm}$; velocity-encoding value, 250 cm/s; retrospective gating) (B) images with corresponding mean velocity–time graph (C) show substantial flow turbulence that starts at level of left ventricular outflow tract (arrow) related to systolic anterior motion of mitral valve leaflets. Through-plane phase-contrast imaging at level of left ventricular outflow tract (LVOT) (TR, 55; acquisition time, 25.5 seconds; voxel size, $2.0 \times 1.2 \times 6 \text{ mm}$; velocity-encoding value, 250 cm/s, retrospectively gated) (solid line, C) and at level of aortic root (TR, 55; acquisition time, 25.6 seconds; voxel size, $2.0 \times 1.2 \times 6 \text{ mm}$; velocity-encoding value, 250 cm/s; retrospective gating) (dashed line, C) show significant flow acceleration at level of LVOT but not at level of aortic valve (AV). Late peaking of LVOT velocity is related to dynamic systolic anterior motion of mitral valve leaflets that increases throughout systole, becoming greatest near end-systole compared with mid peaking aortic valve velocity, which occurs with fixed lesions.