5 Principles of Flow Assessment

Ahmed A. Alsaileek, Fatima Samad,
and A. Jamil Tajik

Introduction

Hemodynamic is concerned with the physical and physiological principles governing the movement of blood through the circulatory system. In other words, hemodynamic is the science and art of the relationship among pressure, viscous resistance to flow, and the volume flow rate in the cardiovascular system. Prior to the application of Doppler echocardiography, hemodynamic assessment was obtained invasively through cardiac catheterization. For most clinical purposes and in daily practice, Doppler echocardiography has replaced cardiac catheterization and becomes the preferred method for hemodynamic assessment. Doppler principle can be applied to calculate flows through different valves, stroke volume, cardiac output, valve areas, regurgitant volumes and shunts. Several animal and clinical studies have validated these methods and have yielded excellent correlations with simultaneously acquired invasive data [1–5].

A. A. Alsaileek
King Saud bin Abdulaziz University for Health Sciences, Riyadh, Saudi Arabia

Adult Cardiology/Cardiac Imaging, King Abdulaziz Cardiac Center, National Guard Health Affairs, Riyadh, Saudi Arabia

F. Samad (✉) · A. J. Tajik
St. Luke's Medical Center, Aurora Cardiovascular Services, Milwaukee, WI, USA

The Doppler Principle

The Doppler principle is applied in echocardiography to enable the determination of the blood flow characteristics such as velocity and direction. In principle, the frequency of the sound waves that are reflected by a stationary object is the same as the frequency of the sound waves transmitted by the transducer. On the other hand, the transmitted frequency of sound waves is altered when reflected by a moving object. If the object is moving toward the transducer, the reflected frequency will be slightly higher and if the object is moving away from the transducer the reflected frequency is slightly lower. This phenomenon is called *Doppler Effect*. In cardiovascular imaging, the moving objects are the red blood cells. The echocardiography instruments determine the *frequency (Doppler) shift* (f D) which is the difference between the transmitted (f T) and the received frequency (f R).

$$f\,\mathrm{D} = f\,\mathrm{R} - f\,\mathrm{T}$$

This Doppler shift is affected by speed of sound in the tissue (c = 1540 m/s), blood flow velocity (V), and the angle (q) between the ultrasound

P. Nihoyannopoulos, J. Kisslo (eds.), *Echocardiography*,
https://doi.org/10.1007/978-3-319-71617-6_5

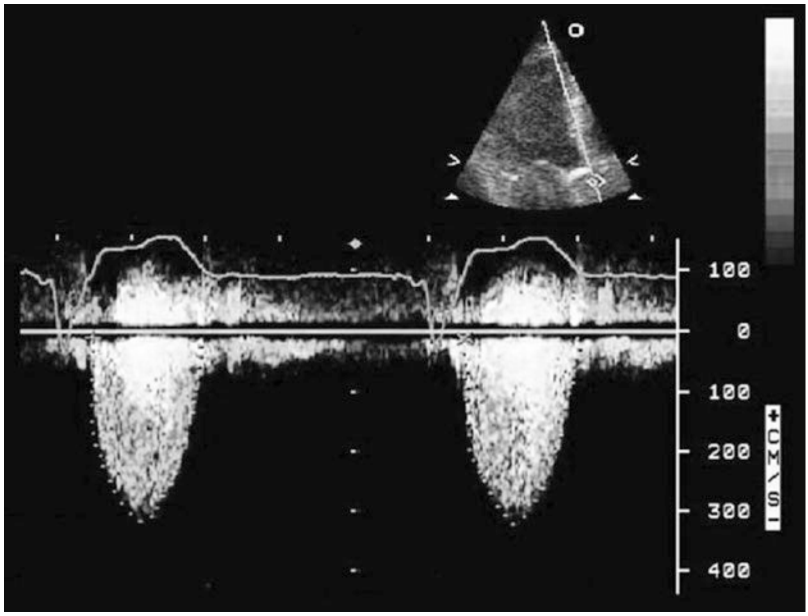

Fig. 5.1 Spectral Doppler analysis, where the time is the horizontal line (baseline), the Y-axis determines the velocity of the flow, and the spectral envelop above and below the baseline determines the flow direction. In this example of aortic stenosis, the transducer is at the apex and the stenotic jet velocities are moving away from the transducer and, therefore, are displayed below the baseline

beam and the blood flow. So the equation can be restated as follows (Doppler equation):

$$f_{\mathrm{D}} = 2 f_{\mathrm{T}} \frac{V \cos\theta}{C}$$

Frequency shift (f D) is, therefore, directly proportional to the blood flow velocity and the equation may be more practically rewritten this way:

$$V = \frac{f_{\mathrm{D}} \mathrm{C}}{2 f_{\mathrm{T}} \cos\theta}$$

(Note that C and f T are constant and the cos q (when the ultrasound beam is parallel to the flow) is equal to one). Therefore, with this equation we can easily determine the frequency shift and hence the blood flow velocity. However, this does not determine the direction of blood flow. The direction can be determined by the echo machine by calculating the frequency shift and giving a positive sign for flow toward the transducer and negative sign for that away from the transducer. In the computer screen this is shown in two ways:

- Spectral Doppler analysis, where the time is the horizontal line (baseline), the *Y*-axis determines the velocity of the flow. The spectral envelops, above and below the baseline, determines the flow direction (Fig. 5.1). In this example the envelop is below the baseline which represent a flow away from the transducer.
- Color-flow Doppler analysis where the flow is color coded according to the direction of the flow. The flow toward the transducer is red and the color away from the transducer is blue (Fig. 5.2).

We will discuss the Doppler assessment of hemodynamics as follows:

- Stroke volume, cardiac output, and cardiac index determination.
- Methodology for calculation of stenotic valve area.
- Principles of calculation of regurgitant volume and effective regurgitant orifice (ERO).
- Determination of transvalvular pressure gradient.
- Estimation of intracardiac pressure and pulmonary artery pressure.

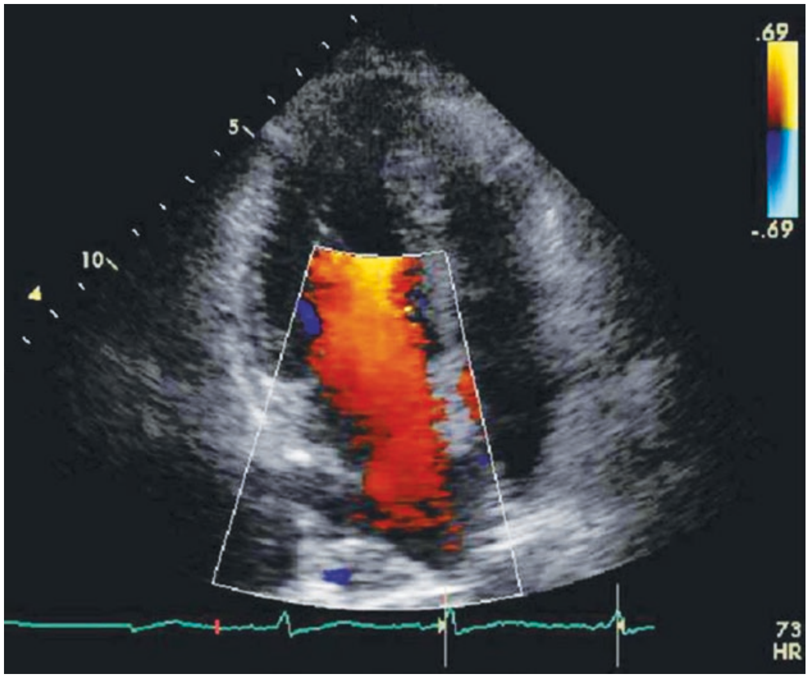

Fig. 5.2 Color Doppler analysis where the flow is color coded according to the direction of the flow. The flow toward the transducer is depicted in shade of *red* and the flow away from the transducer is in shade of *blue*. In this example of the flow across the mitral valve, the transducer is at the apex. The flow is coming toward the transducer, so it is shown as *red*

Four equations or principles are commonly used for these purposes:

1. Hydraulic equation of flow.
2. Continuity equation (law of conservation of mass).
3. Proximal isovelocity surface area (PISA) method.
4. Bernoulli equation.

Principle of Flow Assessment

Blood flow (Q) through a tube, vessel, or across a valve can be derived by a simple hydraulic equation as the product of flow velocity (V) and cross-sectional area (A) of the vessel at the site where velocity is measured. The area can be assumed to be circular and calculated as follows:

$$\begin{aligned}\text{Area} &= \pi r^2 \left(\text{for circular orifice}\right)\\ &= \pi\left(D/2\right)^2 = 3.14\left(D/2\right)^2\\ &= \left(3.14/4\right)D^2 = 0.785D^2\end{aligned}$$

Where D is the diameter of the vessel.

And sometimes the area can be ellipsoid and calculated as follows:

$$\begin{aligned}\text{Area} &= \pi\left(D_1/2 \times D_2/2\right)\left(\text{for ellipsoid orifice}\right)\\ &= 0.785 D_1\, D_2\end{aligned}$$

Flow is the product of area and velocity and can be expressed as:

$$\begin{aligned}&Q\left(\text{cc/s}\right) = A\left(\text{cm}^2\right) \times V\left(\text{cm/s}\right) = 0.785D^2V\\ &\left(\text{for circular orifice}\right) = 0.785\text{D}_1\text{D}_2\text{V}\\ &\qquad\left(\text{for ellipsoid orifice}\right)\end{aligned}$$

This equation assumes a constant flow. However, in pulsatile system (cardiovascular system), flow velocity varies throughout the ejection period and calculation of the volumetric flow is more complex. Therefore, the total flow has to be determined by integrating all individual velocities of the Doppler spectrum over time. In Doppler echocardiography this is known as time velocity

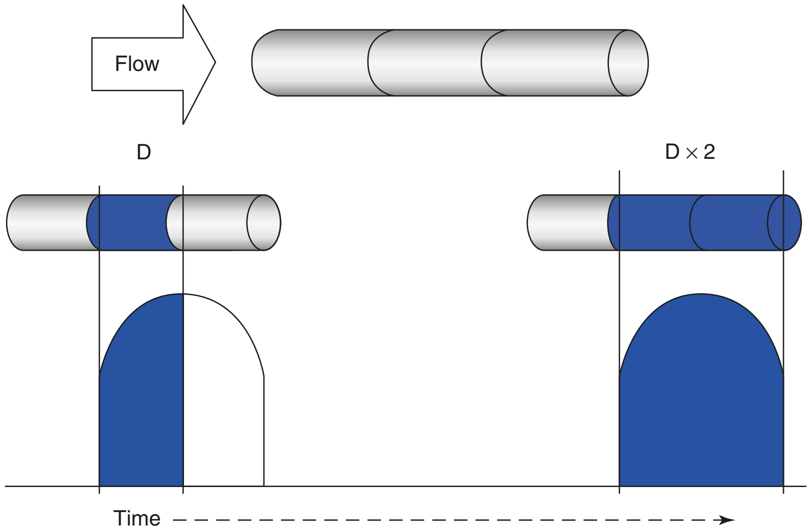

Fig. 5.3 This figure demonstrates the flow in the tube and its representation by spectral method. The relation between the distance (*D*) and time is also shown. The blood travels longer distance with time. In pulsatile system the flow velocity starts at zero, gradually increases to peak, then again decelerates to zero

integral (TVI), which is determined by measuring the area under the curve of the Doppler spectrum. The area under the curve is a measure of the distance the column of fluid travels (Figs. 5.3 and 5.4).

The commercially available echocardiography machines can readily obtain TVI. The modal velocity is derived using the Doppler equation and is dependent on knowing the frequency shift, the velocity of sound in tissue, and the angle between the ultrasound beam and the direction of blood flow. The latter has to be less than 20°; otherwise, significant underestimation of velocity of calculated flow would occur using the assumption of a zero or near-zero intercept between the sound wave and the bloodstream. The area can be planimetered or calculated utilizing two-dimensional echocardiography to obtain diameter (*D*) with a circular ($A = 0.785 \times D$ 2) or elliptic ($A = 0.785$ D 1 D 2) assumption after the diameter(s) (*D*) is measured.

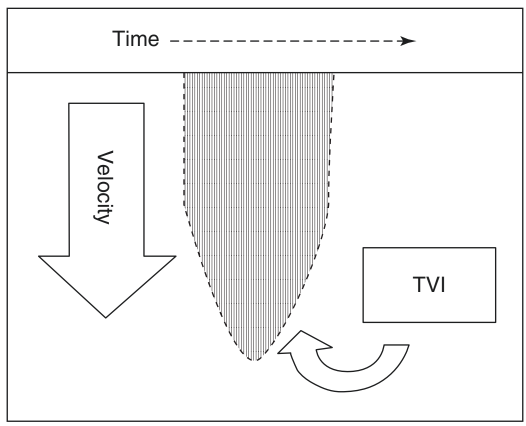

Fig. 5.4 This figure shows schematic illustration of spectral Doppler. The outer border of the spectral Doppler display is traced to determine the TVI

Stroke Volume and Cardiac Output

One of the fundamental functions of the heart as a pump is to provide adequate amount of blood flow for the normal function of the human body. In the past, the cardiac output determination required invasive comprehensive and time-consuming methods based upon Fick and indicator dilution principles. Nowadays, stroke volume and cardiac output can be noninvasively and reliably measured by 2D echo/Doppler methodology.

The forward stroke volume can be determined by the Doppler method mentioned earlier, which can basically be applied to any of the cardiac valves assuming no significant valvular regurgitation is present. Usually the flow through the aortic valve (or left ventricular outflow tract—LVOT) is calculated because the aortic annulus has the least change in size during the cardiac cycle [4, 6].

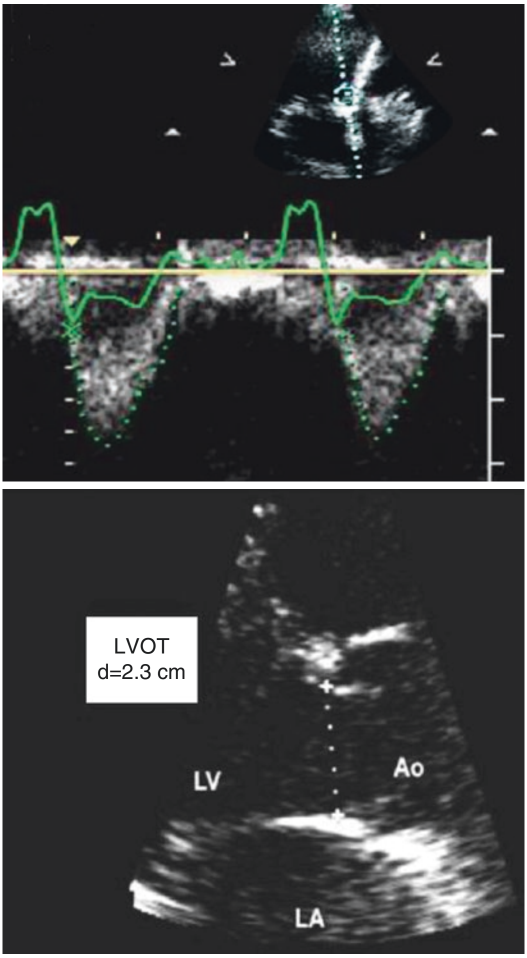

Fig. 5.5 This figure illustrates the Doppler method to determine stroke volume. The diameter of the aortic annulus is usually obtained from parasternal long-axis view (lower). In this example aortic annulus diameter is equal to 2.3 cm. The maximum diameter should be recorded. The TVI of the aortic annulus is normally obtained from apical long-axis view and occasionally utilizing 5-chamber view (upper). The TVI of LVOT is obtained by tracing the outer border of PW Doppler signal of LVOT. In this example average TVI is equal to 20.0 cm

The aortic annulus diameter is measured in the parasternal long-axis view (Fig. 5.5 -lower) and the maximum diameter is obtained. The velocity at the annulus is obtained utilizing the apical long-axis view and by placing the pulsed-wave (PW) Doppler sample volume at the level of aortic annulus. The closing click of the aortic valve should be recognized to insure good position. This velocity is then planimetered along its outer edge to obtain the TVI (Fig. 5.5 -upper). If significant aortic regurgitation is present, the flow through this valve cannot be used to calculate the cardiac output. Errors can be incurred in the area calculation, mostly related to the diameter measurement (the errorin area calculation is roughly doubled compared to the error in diameter measurement), but also includes inappropriate geometric assumptions and data acquisition at a different site from where velocity was recorded. Looking at the numbers in the figures (Fig. 5.5) we can calculate the stroke volume as follows:

$$\text{Stroke volume}(\text{SV}) = 0.785 \times \text{D}^2 \times \text{TVI} = 0.785 \times 2.3^2 \times 20.0 = 83\text{ml}$$

Cardiac output is the product of stroke volume and heart rate (assume the heart rate is 65 beats/min).

Cardiac output (CO) = SV × HR = 83 × 65 = 5398 ml/min = 5.40 l/min (by converting the milliliter to liter).

Cardiac index is obtained by dividing the cardiac output by body surface area (BSA).

$$\text{Cardiac index}(\text{CI}) = \text{CO} / \text{BSA}(\textit{liter} / \text{min}/\text{m2})$$

Alternatively, but uncommonly, cardiac output can be obtained from pulmonic valve or right ventricular outflow tract. In the short axis at the level of aortic valve the ultrasound beam is almost parallel to the blood flow at the level of pulmonic valve or right ventricular outflow tract. The TVI is obtained by pulsed-wave Doppler at the same level where the right ventricular outflow tract area is measured in 2D echocardiography [7]. The cardiac output can be obtained at any valve assuming there is no regurgitation of the valve so the cardiac output is constant for any given cardiac cycle.

Continuity Equation

Continuity equation is based on the principle of conservation of mass. This principle states that, under conditions of cardiovascular stability

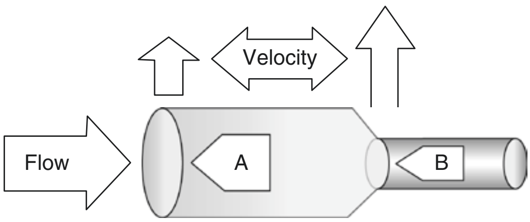

Fig. 5.6 This figure shows two tubes with different diameter in continuity. The continuity equation states that the flow volume at point A is equal to that of point B and mathematically written as follows: (Area A × Velocity A = Area B × Velocity B). Note that the velocity is higher with smaller area

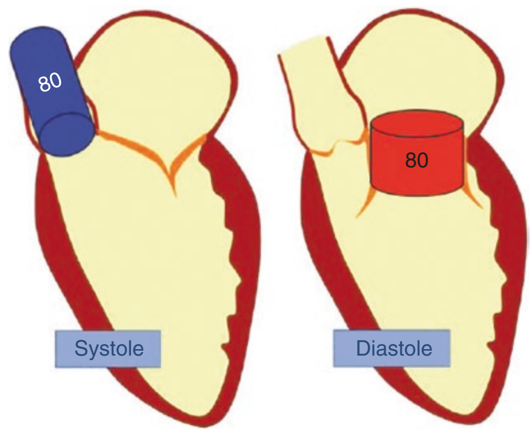

Fig. 5.7 This figure demonstrates the continuity between mitral and aortic valves' stroke volume. The volume of blood that comes in through mitral valve must go out through aortic valve

without any regurgitation or shunt, the net blood volume at any part of circulation must equal the net blood volume at any other part next to it (what comes in must go out) (Fig. 5.6). This situation is true under certain assumptions:

- The two points are directly connected.
- Blood is neither added nor removed from the system.

Clinical Applications of Continuity Equation

In applying the continuity equation to the blood flow in and out of the heart, the mitral valve stroke volume is equal to aortic valve stroke volume provided there is no mitral or aortic regurgitation (Fig. 5.7). In the same way the aortic stroke volume is equal to pulmonic valve stroke volume provided no significant regurgitation is present in any of the valves, and no intracardiac shunt is present. The tricuspid valve stroke is not commonly used in clinical practice because of complex geometry for valve area calculation and because of the presence of tricuspid regurgitation in the majority of normal subjects. The aortic valve stroke volume is calculated by measuring the aortic annulus diameter in parasternal long-axis view at maximum valve opening in systole. The TVI is obtained utilizing pulsed-wave Doppler sample volume at the level of aortic annulus on the apical long-axis or 5-chamber view. The aortic valve area (AVA) is assumed to be circular and, hence, the area is calculated from the obtained aortic annulus diameter (*D* AA).

$$\text{AVA} = \pi\left(D_{\text{AA}}/2\right)^2 = 0.785\text{D}_{\text{AA}}{}^2$$

The stroke volume at the level of aortic valve is then calculated as follows:

$$\text{SV}_{\text{AA}} = \text{AVA} \times \text{TVI}_{\text{AA}}$$
$$\text{SV}_{\text{AA}} = 0.785 \times D_{\text{AA}}{}^2 \times \text{TVI}_{\text{AA}}$$

In calculating the mitral valve stroke volume, the mitral valve area is calculated either by assuming a circular or ellipsoid geometry. For practical purposes, the circular shape is the most often used. The mitral annulus diameter (*D* MA) is obtained from the apical 4-chamber view at maximum valve opening in diastole. The TVI is obtained from the same view by pulsed-wave Doppler sample volume at the level of the mitral annulus.

$$\text{MVA} = 0.785 \times D_{\text{MA}}{}^2$$

And therefore:

$$SV_{MA} = MVA \times TVI_{MA}$$
$$SV_{MA} = 0.785 \times D_{MA}^{\ 2} \times TVI_{MA}$$

Mathematically the continuity equation means that stroke volume through mitral valve is equal to the stroke volume through aortic valve:

$$TVI\,1 \times Area\,1(mitral) = TVI\,2 \times Area\,2(aortic)$$
$$0.785 \times D_{MA}^{\ 2} \times TVI_{MA} = 0.785 \times D_{AA}^{\ 2} \times TVI_{AA}$$

The concept of continuity is very useful clinically to assess mitral or aortic regurgitant volume, regurgitant fraction, regurgitant orifice, as well as the stenotic mitral or aortic valve area.

Mitral Regurgitation

In mitral regurgitation (MR) (Figs. 5.8 and 5.9), the aortic stroke volume is less than the forward mitral stroke volume because part of the blood contained in the LV at the end of diastole is ejected back to the left atrium through the regurgitant mitral valve.

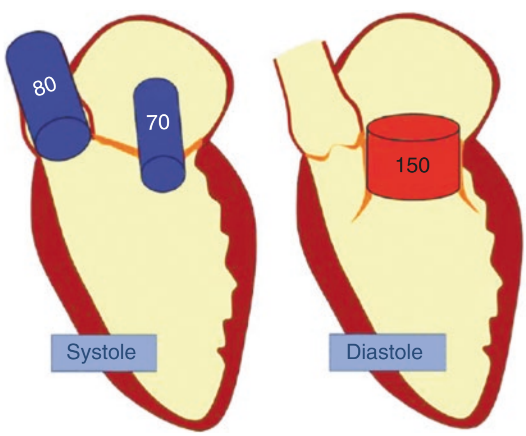

Fig. 5.8 This figure shows schematic illustration of the principle of calculating mitral regurgitant volume based on hydraulic formula. Mitral inflow volume of 150 cc (what goes in) and aortic stroke volume of 80 cc (what goes out) are calculated based on the continuity equation. The difference (150 − 80 = 70 cc) must represent the mitral regurgitation volume

$$\text{Aortic stroke volume} = \text{mitral stroke volume} - RV(\text{Mitral})$$

Therefore,

$$\text{Mitral RV} = \text{mitral stroke volume} - \text{Aortic stroke volume}$$
$$= \left(0.785 \times D_{MA}^{\ 2} \times TVI_{MA}\right) - \left(0.785 \times D_{AA}^{\ 2} \times TVI_{AA}\right)$$

Aortic Regurgitation

In aortic regurgitation, the aortic stroke volume is more than the mitral stroke volume because the regurgitant volume through aortic valve will be added to the subsequent stroke volume from mitral valve (Figs. 5.9 and 5.10). The same methodology described to assess the mitral valve regurgitant volume is used to assess the aortic regurgitant volume. The difference is that the aortic stroke volume will be higher than that of mitral stroke volume (see earlier).

$$\text{Aortic \textbackslash stroke \textbackslash volume} = \text{mitral \textbackslash stroke \textbackslash volume} + RV(\text{Aortic})$$

Therefore,

$$\text{Aortic RV} = \text{Aortic stroke volume} - \text{mitral stroke volume} = \left(0.785 \times D_{AA}^{\ 2} \times TVI_{AA}\right) - \left(0.785 \times D_{MA}^{\ 2} \times TVI_{MA}\right)$$

Mitral Stenosis

The mitral valve area in the presence of mitral stenosis can also be estimated using the continuity equation. The stroke volume at mitral valve orifice is equal to the aortic stroke volume (the

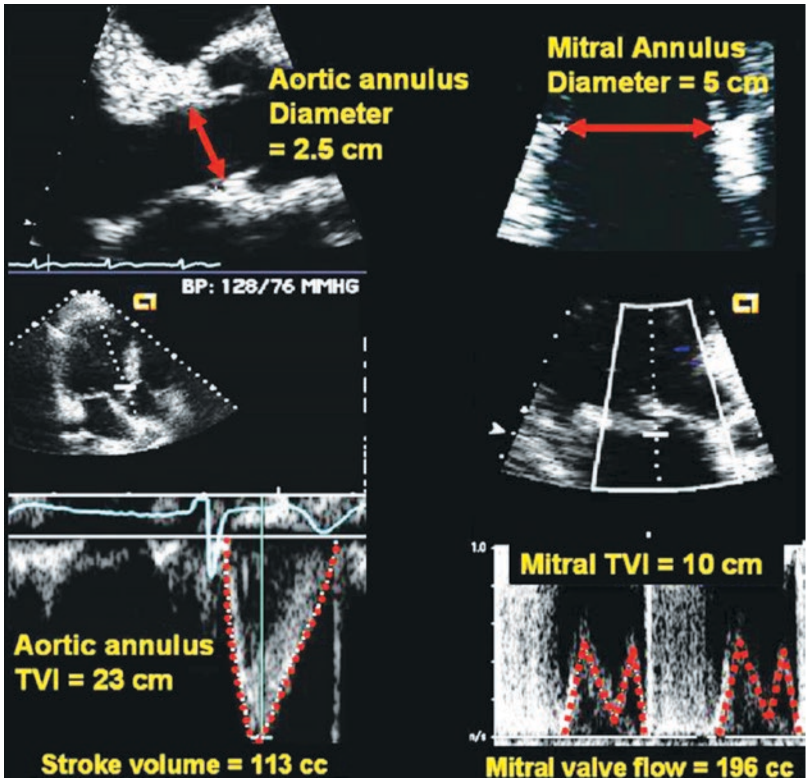

Fig. 5.9 The measurement needed for continuity equation in mitral regurgitation and aortic regurgitation. The mitral inflow volume is calculated from the mitral inflow TVI and mitral annulus diameter (0.785 *D* MA 2 × TVI MA = 196 cc). The aortic stroke volume is calculated from the aortic TVI and aortic annulus diameter (0.785 × *D* AA 2 × TVI AA = 113 cc). The difference (196 − 113 = 83 cc) must be the MR volume

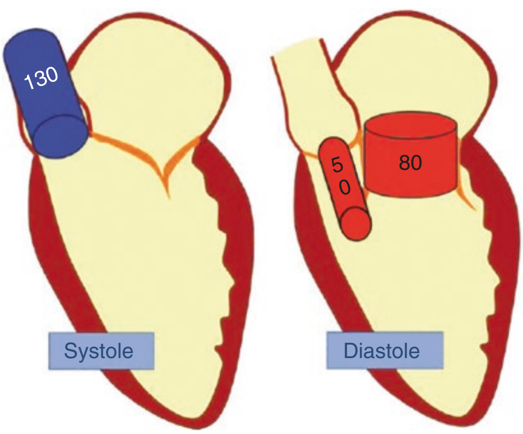

Fig. 5.10 This figure shows schematic illustration of the principle of calculating aortic regurgitant volume based of hydraulic formula. Mitral inflow volume 80 cc (what goes in) and aortic stroke volume 130 cc (what goes out) are calculated based on the continuity equation. The difference (130 − 80 = 50 cc) must represent the aortic regurgitation volume

method to obtain the aortic SV is already discussed). The aortic annulus area multiplied by the aortic TVI is equal to MVA multiplied by the mitral TVI. Mitral valve TVI is obtained by continuous-wave Doppler across the mitral valve from apical 4-chamber or apical long-axis views. The MVA can then be calculated as follows:

$$\text{MVA} \times \text{flow} = \text{AVA} \times \text{flow}$$
$$\text{MVA} \times \text{TVA}_{\text{MA}} = 0.785 \times D_{\text{AA}}{}^{2} \times \text{TVI}_{\text{AA}}$$
$$\text{MVA} = 0.785 \times D_{\text{AA}}{}^{2} \times \text{TVI}_{\text{AA}} / \text{TVI}_{\text{MA}}$$

The limitation of this method is the presence of more than trivial/mild MR, which may be present in a significant number of patients [8, 9].

Aortic Stenosis

The same principle can be applied to any two orifices connected in series or tube at two different points provided that all the flow goes in one direction. This can be applied to assess the aortic valve area in aortic stenosis (*what comes in must go out*) [10, 11] (Fig. 5.11). The volume of blood going through LVOT (proximal) should be equal to that going through a stenotic aortic valve

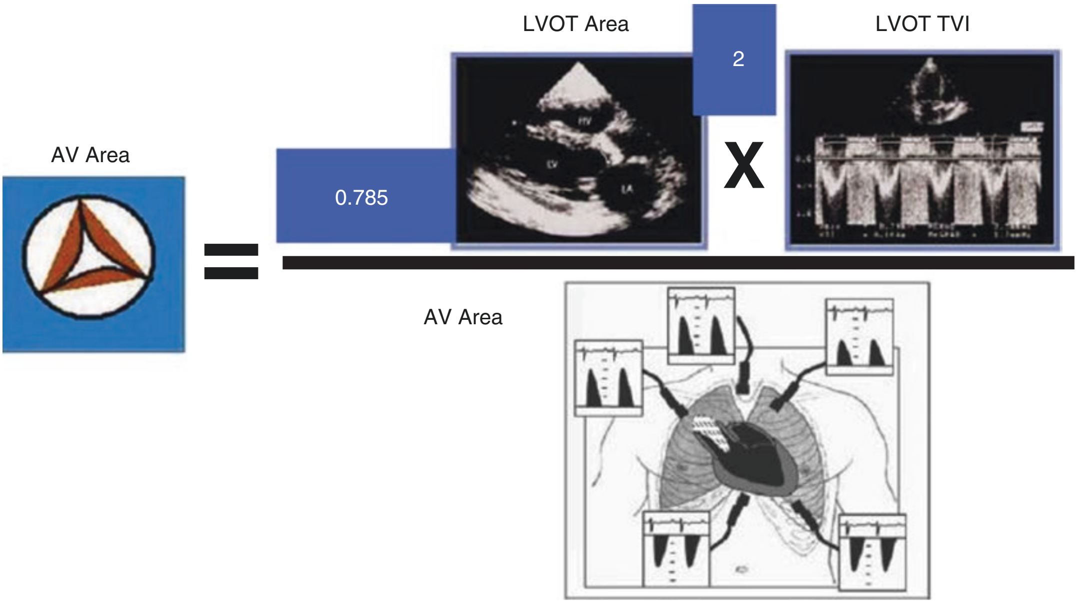

Fig. 5.11 This figure shows the continuity method to assess AS severity. The flow rates at the LVOT and AV are the same. The LVOT flow rate is calculated from the LVOT TVI and diameter (see also Fig. 5.5). The TVI of AV is obtained by tracing the outer border of CW Doppler signal of AV. Then the AVA is calculated as: AVA (cm^2) = (0.785 D LVOT 2 × TVI LVOT)/TVI AA

(distal). The LVOT diameter and TVI (TVI LVOT) is determined as discussed previously. One important point to make is to avoid the area of flow acceleration (due to severe aortic stenosis) in the LVOT TVI determination by moving the sample volume about 1 cm below the aortic annulus. The aortic TVI (TVI AA) is obtained by CW Doppler across the aortic valve. This should be done from multiple locations (Fig. 5.11) and the highest velocity is used in the calculation. Aortic valve area (AVA) is calculated as:

$$\mathrm{AVA}\left(\mathrm{cm}^2\right)\times \mathrm{TVI}_{\mathrm{AA}} = \left(0.785\times D_{\mathrm{LVOT}}{}^{2}\times \mathrm{TVI}_{\mathrm{LVOT}}\right)$$

And then,

$$\mathrm{AVA}\left(\mathrm{cm}^2\right) = 0.785\times D_{\mathrm{LVOT}}{}^{2}\times \mathrm{TVI}_{\mathrm{LVOT}} / \mathrm{TVI}_{\mathrm{AA}}$$

Echocardiographic findings of patients with tricuspid regurgitation generally mirror those found in MR. Therefore, the tricuspid valve stroke volume can be compared to the pulmonic valve stroke volume and the regurgitant volume can be estimated, but clinically this method is less often used than the PISA method [12].

Proximal Isovelocity Surface Area Method

Proximal isovelocity surface area (PISA) method is based on the law of conservation of flow. The law states the flow rate at two consecutive points is identical. The method uses the advantage of aliasing in color Doppler imaging. In color Doppler the flow is given red or blue color if the direction of flow is toward and away from the transducer, respectively. As the flow approaches a narrowed orifice (stenotic or regurgitant), its velocity increases, in a shape of isovelocity hemispheric shell as blood flow converges from all directions toward the orifice (Fig. 5.12). The flow rate at the surface of the hemispheric shell is equal to the flow rate at the regurgitant orifice (law of conservation of the flow). As the flow converges toward the orifice, it accelerates and aliasing occurs if the velocity exceeds the Nyquist limit. This can be nicely shown using the color Doppler imaging where the color changes from red to blue in a nice hemisphere. If the Nyquist limit is

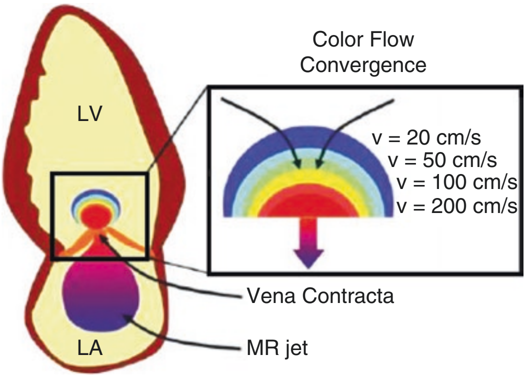

Fig. 5.12 Schematic illustration of the PISA. The flow accelerates as it approaches a narrow orifice forming hemispheric shell. This can be shown on color Doppler (as in case of MR). The flow converges as it approaches the regurgitant mitral valve. The color changes due to high velocity exceeding the Nyquist limit and forming hemispheric shell (PISA)

decreased, the aliasing occurs with lower velocity and starts further away from the regurgitant orifice making the hemisphere (PISA) larger. Using this principle, flow rate at the surface of the hemispheric shape can be estimated. The flow rate can be calculated by multiplying the area by velocity (Flow rate = Area × Velocity). The area in case of valve is circular so it is calculated as (*p r* 2), but in case of hemisphere surface area is calculated as ($2\ \pi r^2$ or $6.28\ r^2$), the *r* is the radius of the hemisphere. The radius of the hemisphere can be measured from the surface to the narrowest area of color flow, which is closely related to the regurgitant or stenotic orifice. The flow rate at the hemisphere (PISA) surface is Flow rate $= AV = 2\pi r^2 V_{\text{aliasing}} = 6.28 r^2 V_{\text{aliasing}}$.

Clinical Applications of PISA Method

Mitral Regurgitation

In MR, the apical 4-chamber view with color Doppler is commonly used (Figs. 5.12 and 5.13). However, parasternal long-axis view is sometimes used in case of eccentric jet where the PISA is better visualized. The Nyquist limit is shifted downward in the direction of the flow of MR jet. The velocity of the PISA is measured at midlate systole (the same time the maximum MR velocity occurs) [13, 14].

$$\text{Area}(\text{ERO}) \times V_{\max}\text{MR} = 6.28 r^2 V_{\text{aliasing}}$$

$$\text{Effective regurgitant orifice}(\text{ERO}) = 6.28 r^2 V_{\text{aliasing}} \,/\, V_{\max}\text{MR}$$

As discussed in the calculation of the stroke volume, it is the product of area and TVI. The same method can be applied to calculate the regurgitant volume (RV). The area here is known (see earlier), which is the ERO of the mitral valve. The TVI is nothing but the TVI of the MR jet. Therefore:

$$\text{Mitrla RV} = \text{ERO}(\text{PISA}) \times \text{TVI}_{\text{MR}}(\text{CW\textbackslash Doppler}) = \left(6.28 r^2 V_{\text{aliasing}} / V_{\max}\text{MR}\right)\text{TVI}_{\text{MR}}$$

Width of vena contracta is another quantitative method to assess MR. It is defined as the narrowest cross-sectional area of a jet. This can be easily seen, while obtaining the zone of flow convergence, above the mitral valve on the left atrial side (Fig. 5.12). On transthoracic color-flow mapping it has been shown that vena contracta width predicts angiographic severity of MR [15]. Compared to continuity equation, vena contracta of more than 5 mm correlates well with severe MR [16].

Semiquantitative Doppler methods to evaluate the MR are less sensitive and are considered complementary in MR evaluation.

Color-flow Doppler: The features of severe MR seen on color-flow Doppler imaging arise from the high-energy transfer of a large volume of blood into the left atrium, producing "jets" in the left atrium. Color-flow Doppler remains the easiest and the best method to screen for MR. It also provides semiquantitative assessment of the MR severity. The ratio of the MR jet area to the total left atrial area has been reported to correlate well with MR severity [17]. Severe MR is characterized by large jet (>40%) and extending into the pulmonary veins. However, jets are very sensitive to

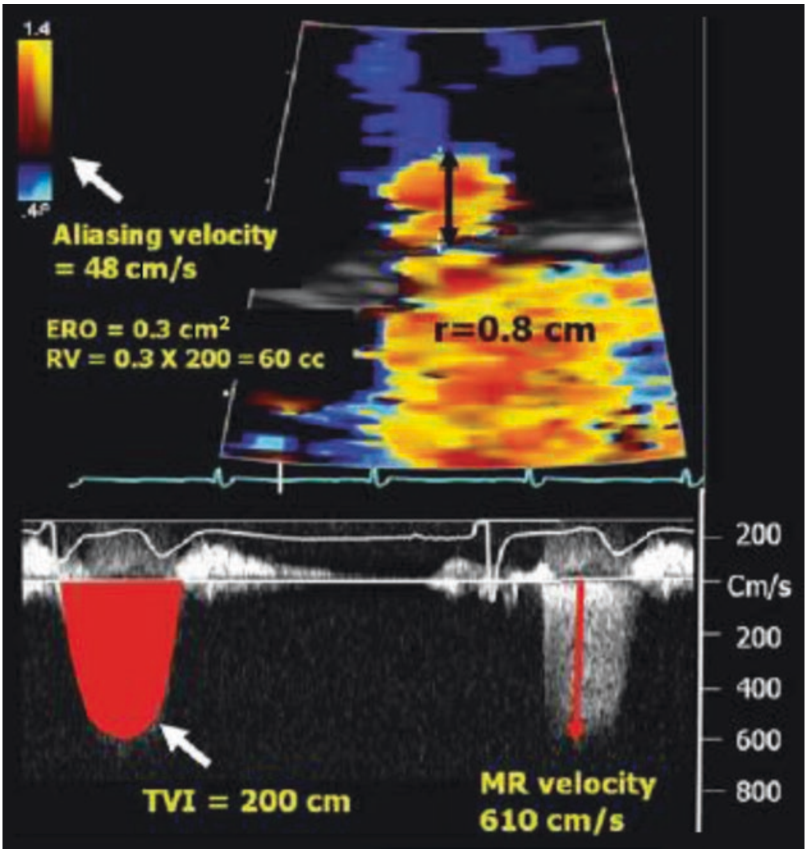

Fig. 5.13 This figure shows the PISA methods to evaluate the severity of MR. The PISA is obtained from the apical 4-chamber view. The baseline is shifted downward toward the regurgitant jet flow (*arrow*). Then the PISA diameter is measured from the surface of the hemisphere to the narrowest area on color Doppler (0.8 cm). The CW Doppler is utilized to measure the mitral regurgitation TVI and peak velocity. The effective regurgitant orifice (ERO) and the regurgitant volume (RV) are calculated as follows: ERO = 6.28 r^2 V aliasing/V max MR = 6.28 × .82 × 48/610 = 0.3 cm^2 and RV = TVI × ERO = 200 × 0.3 = 60 cc

instrument settings, and the size of a color jet may be misleading such as in an eccentric jet; thus, reliance on these size judgments alone may not be sound practice [18–20].

Doppler of pulmonary veins: Doppler interrogation of the pulmonary veins has produced insights into hemodynamics. In MR evaluation, pulsed-wave Doppler of the left and right upper pulmonary veins is performed from the apical 4-chamber view. In hemodynamically severe MR, the flow in one or more pulmonary veins will show systolic flow reversal. This echo feature is the analog of the V wave seen in the left atrial pressure tracing and on a pulmonary artery wedge pressure tracing.

Continuous-wave Doppler of the MR jet: If the flow signal can be aligned parallel to the beam, in severe MR, the Spectral Doppler of the jet will appear uniformly dense throughout its duration and have a well-defined envelop. MR jet velocity does not correlate with the severity. Other features that are associated with severe MR include dilated or dysfunctional left ventricle, dilated left atrium, elevated pulmonary artery systolic pressure, and the presence of significant tricuspid regurgitation. In severe decompensated MR, the tricuspid regurgitation peak velocity will be increased as the result of pulmonary hypertension.

D PISA in some studies has proved to be superior to 2D measures for distinguishing moderate from severe MR, especially for eccentric and asymmetric jets.

Tricuspid Regurgitation

As with MR, severity of tricuspid regurgitation (TR) can be estimated by effective regurgitant orifice (ERO) by measuring the size of proximal flow convergence zones (PISA) [21].

$$\text{Area}(\text{ERO}) \times V_{\max}\text{TR} = 6.28\text{r}^2 V_{\text{aliasing}}$$
$$\text{ERO}(\text{Tricuspid}) = 6.28 r^2 V_{\text{aliasing}} / V_{\max}\text{TR}$$

And even the regurgitant volume (RV) can be calculated:

$$\text{RV}(\text{Tricuspid}) = \text{Area}(\text{ERO}) \times \text{TVI} = \left(6.28\text{r}^2 V_{\text{aliasing}} / V_{\max}\text{TR}\right) \times \text{TVI}_{\text{TR}}$$

Aortic Regurgitation

The severity of aortic regurgitation can be quantitatively assessed utilizing the PISA method [22]. The apical 5-chamber and apical long-axis views are used to visualize the PISA. However, parasternal long-axis view is sometimes used in case of eccentric jet where the PISA is better visualized. The Nyquist limit baseline is shifted toward the flow of the aortic regurgitation jet (Fig. 5.14). The PISA radius is measured at early diastole [23]. The maximum aortic regurgitation (AR) velocity in early diastole is measured. The ERO is calculated as follows:

$$\textbf{Aortic ERO} = \textbf{PISA flow rate} / \textbf{AR regurgitant velocity} = \mathbf{6.28r^2}\left(\mathbf{V_{aliasing}} / \textbf{AR regurgitant velocity}\right)$$

The aortic regurgitant volume is calculated by tracing the aortic regurgitation jet outer surface to

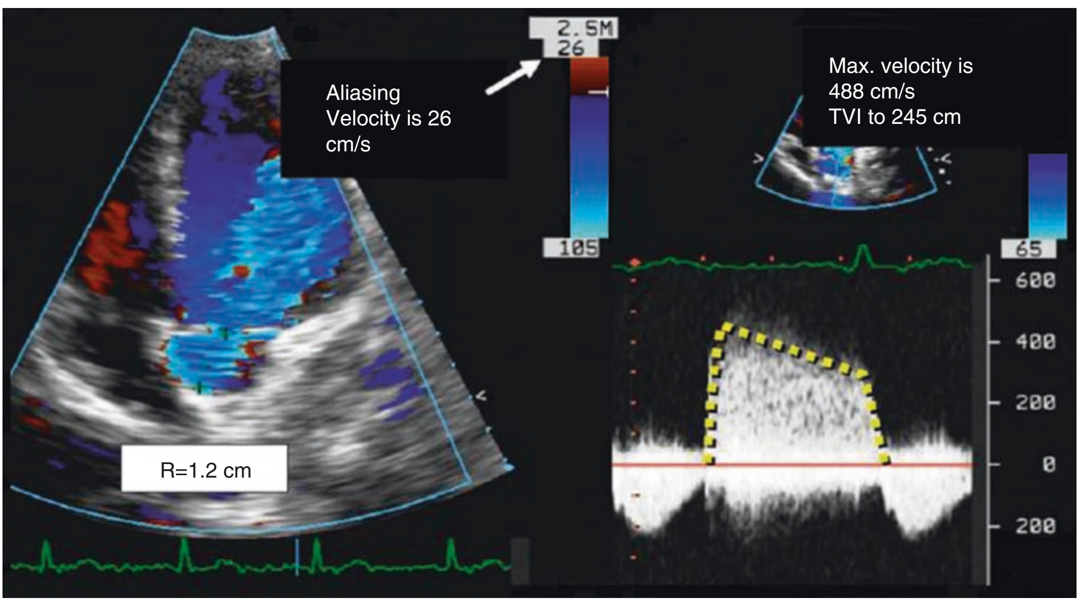

Fig. 5.14 This figure shows the PISA method for calculation of the ERO in case of aortic regurgitation. The PISA is obtained from the apical long-axis or 5-chamber view. The baseline is shifted upward toward the regurgitant jet flow (*arrow* regurgitant volume). Then the PISA diameter is measured from the surface of the hemisphere to the narrowest area on color Doppler (1.2 cm). The CW Doppler is utilized to measure the aortic regurgitation TVI and peak velocity. The effective regurgitant orifice (ERO) and the regurgitant volume (RV) are calculated as follows: ERO = 6.28r2 Valiasing/Vmax AR = 6.28 × 1.2 × 1.2 × 26/488 = 0.5 cm^2 and RV = TVI × ERO = 245 × 0.5 = 118 cc

measure the aortic regurgitation TVI (TVIAR); therefore:

$$\text{flow volume}(\text{RV}) = \text{Area}(\text{ERO}) \times \text{TVI} = \left(6.28 r^2 V_{\text{aliasing}} / V_{\text{max}} \text{AR}\right) \times \text{TVI}_{\text{AR}}$$

Vena Contracta It is the narrowest portion of the jet crossing the plane of the valve. It lies immediately next to the area of flow convergence. It can be measured from parasternal long-axis or apical long axis views. To optimize the visualization, the echo sector should be the narrowest possible and the depth decreased. Vena contracta of equal to or more than 6 mm has been shown to correlate with severe aortic regurgitation. This method was found useful even in eccentric jet [24–26]. Some investigators [27] have used the cross-sectional area of the vena contracta to evaluate aortic regurgitation severity. The approach sounds promising and 3D echo may help to further tune this approach [24].

There are several Doppler methods to evaluate aortic regurgitation severity, most of which are less accurate and should not be used alone to guide clinical decisions [25–27].

Regurgitant Jet Size Similar to most of the valvular lesions, color-flow Doppler is the initial screening method for the aortic regurgitation. With the advent of color Doppler many investigators had attempted to use the regurgitant jet size to quantify aortic regurgitation severity. However, this method has suffered several limitations [28, 29]. The jet is frequently eccentric and appears much smaller even in the presence of significant aortic regurgitation, and its spread may be affected by the shape of the ventricular septum. Thus the size of the jet is only used as a screening tool after which other quantitative methods are used.

Jet Height to LVOT Height Ratio This method depends on using the color Doppler in parasternal long-axis view to record the jet height at the valve orifice and compare it to the LVOT height (Fig. 5.15).

This is related to the regurgitant orifice. The higher the ratio, the more severe is the aortic regurgitation. This method also has limitations. Eccentric jet may be difficult to assess. The shape of the regurgitant orifice affects the height of the jet on color Doppler images. Due to 2-Dimensional nature of the color-flow Doppler alignment on 2-Dimentional echo, they may not be perfectly aligned as they may not be in the same plane [30]. *Pressure half-time (PHT)*—Aortic regurgitation jet PHT is measured using continuous-

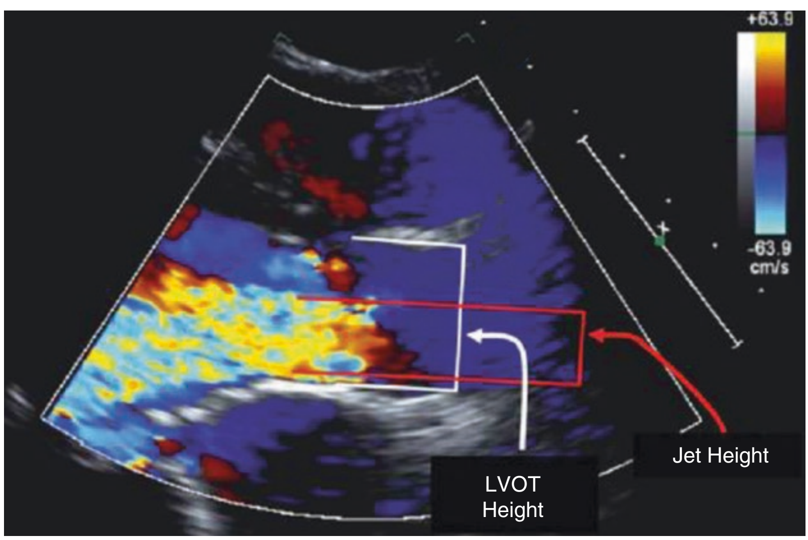

Fig. 5.15 Transthoracic echocardiogram at long-axis view demonstrates color Doppler method of the ratio of jet height to LVOT height to assess the severity of aortic regurgitation

wave Doppler spectral display from apical long-axis view. It represents how quickly the aortoventricular pressure gradient equalizes during diastole (Fig. 5.16) [31]. The larger the regurgitant orifice, the more quickly the pressure equalizes and the velocity falls. A group of investigators has demonstrated that this sign is useful to judge about severe aortic regurgitation. The PHT of less than 200 ms has been used to indicate severe aortic regurgitation [32]. However, it is important to keep in mind that numerous other factors can impact this, such as systemic vascular resistance and ventricular compliance [33].

Aortic Diastolic Flow Reversal One of the earliest Doppler techniques to study aortic regurgitation was to examine retrograde diastolic flow reversal in the ascending and descending aorta or arch. Different numbers had been proposed to the measurement of the flow reversal TVI trying to assess severity. In general, flow reversal throughout the entire diastole is consistent with significant aortic regurgitation [34].

Mitral Stenosis

The PISA method can also be used to estimate mitral stenosis (MS) [35, 36]. Because of the stenotic mitral valve, blood flow accelerates at the left atrial side and converges forming nice PISA (Fig. 5.17). With the appropriate setting (moving the Nyquist limit baseline upward the same direction of flow while interrogating the mitral valve from apical views), PISA can be optimized. The flow rate at the PISA surface is equal to the flow rate at the mitral valve orifice. The mitral valve maximum velocity is obtained by continuous Doppler through mitral valve at early diastole. Then the stenotic area (MVA) can be calculated as follows:

$$\text{MVA} = 6.28\text{r}^2\ V\text{aliasing/stenotic MV max velocity}$$

The truly hemispheric shells occurs if surface of the valve is flat with the leaflets apposed at 180°. This PISA method in case of mitral stenosis is not perfect because mitral valve at maximum opening is not fl at (less than 180°) (Fig. 5.17). So the angle (alpha = *a*) has to be corrected for, by dividing this angle by 180°. The estimation of the angle is crude approach and a

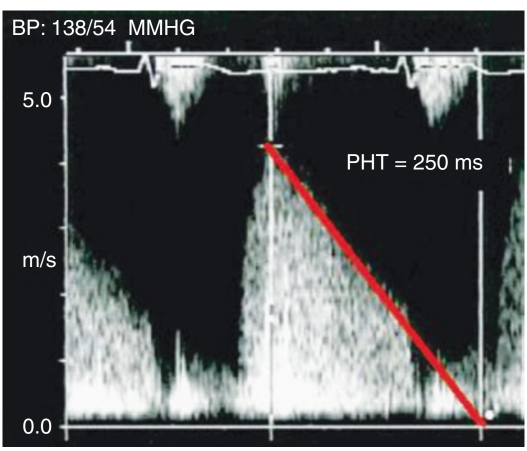

Fig. 5.16 CW spectral Doppler of the aortic regurgitation showing the slope of pressure gradient decays. The pressure halftime is then calculated to assess the severity of aortic regurgitation

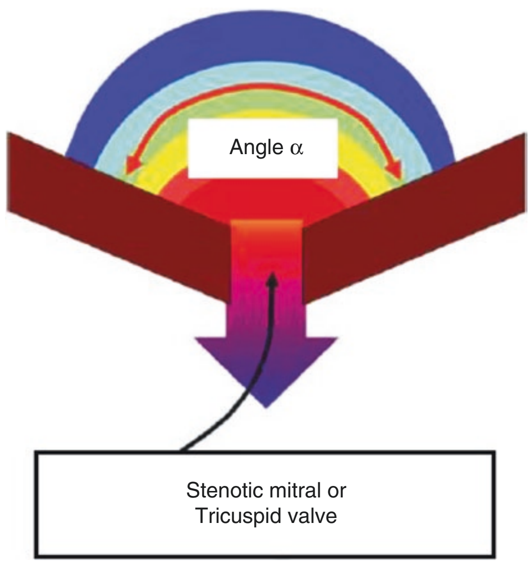

Fig. 5.17 This figure shows the PISA at the mitral or tricuspid valves in case of mitral or tricuspid stenosis. Occasionally, the valve leaflets form an angle while opening during diastole (not flat surface). This angle (*a*) is less than 180 and needs correction factor (*a*/180) when using the PISA to calculate stenotic lesion. This makes the method less perfect than for the regurgitant lesions

significant source of error [35–37]. MVA then calculated as:

$$\mathrm{MVA} = 6.28\mathrm{r}^2 \times V\mathrm{aliasing} / \text{stenotic MV maximum velocity}\left(a/180\right)$$

The advantage of this approach is that it is not affected by coexisting mitral or aortic regurgitation [38]. The tricuspid stenosis is not common. The severity can be quantitatively estimated in an identical way as the mitral stensos. Utilizing PISA, the flow at the PISA surface is equal to the flow rate at the tricuspid valve orifice (TVA). Because of complex valve geometry similar to the mitral stenosis, the angle needs to be corrected for (a /180).

$$\mathrm{TVA} = 6.28\mathrm{r}^2\, V\mathrm{aliasing} / \text{stenotic TV maximum velocity} \times \left(a/180\right)$$

Again this approach is not perfect and less commonly used than other methods such as mean gradient and pressure half-time.

References

1. Callahan MJ, Tajik AJ, Su-Fan Q, Bove AA. Validation of instantaneous pressure gradients measured by continuous-wave Doppler in experimentally induced aortic stenosis. Am J Cardiol. 1985;56(15):989–93.
2. Currie PJ, Seward JB, Chan KL, Fyfe DA, Hagler DJ, Mair DD, et al. Continuous wave Doppler determination of right ventricular pressure: a simultaneous Doppler-catheterization study in 127 patients. J Am Coll Cardiol. 1985;6(4):750–6.
3. Burstow DJ, Nishimura RA, Bailey KR, Reeder GS, Holmes DR Jr, Seward JB, Tajik AJ. Continuous wave Doppler echocardiographic measurement of prosthetic valve gradients. A simultaneous Doppler-catheter correlative study. Circulation. 1989;80(3):504–14.
4. Lewis JF, Kuo LC, Nelson JG, Limacher MC, Quinones MA. Pulsed Doppler echocardiographic determination of stroke volume and cardiac output: clinical validation of two new methods using the apical window. Circulation. 1984;70:425–31.
5. Stewart WJ, Jiang L, Mich R, Pandian N, Guerrero JL, Weyman AE. Variable effects of changes in flow rate through the aortic, pulmonary and mitral valves on valve area and flow velocity: impact on quantitative Doppler flow calculations. J Am Coll Cardiol. 1985;6:653–62.
6. Zoghbi WA, Quinones MA. Determination of cardiac output by Doppler echocardiography: a critical appraisal. Herz. 1986;11(5):258–68.
7. Maslow A, Comunale ME, Haering JM, Watkins J. Pulsed wave Doppler measurement of cardiac output from the right ventricular outflow tract. Anesth Analg. 1996;83:466–71.
8. Karp K, Teien D, Eriksson P. Doppler echocardiographic assessment of the valve area in patients with atrioventricular valve stenosis by application of the continuity equation. J Intern Med. 1989;225:261–6.
9. Nakatani S, Masuyama T, Kodama K, Kitabatake A, Fujii K, Kamada T, et al. Value and limitations of Doppler echocardiography in the quantification of stenotic mitral valve area: comparison of the pressure half-time and the continuity equation methods. Circulation. 1988;77:78–85.
10. Otto CM, Pearlman AS, Gardner CL, Enomoto DM, Togo T, Tsuboi H, Ivey TD. Experimental validation of Doppler echocardiographic measurement of volume flow through the stenotic aortic valve. Circulation. 1988;78:435–41.
11. Skjaerpe T, Hegrenaes L, Hatle L. Noninvasive estimation of valve area in patients with aortic stenosis by Doppler ultrasound and two-dimensional echocardiography. Circulation. 1985;72:810–8.
12. Miyatake K, Okamoto M, Kinoshita N, Ohta M, Kozuka T, Sakakibara H, Nimura Y. Evaluation of tricuspid regurgitation by pulsed Doppler and two-dimensional echocardiography. Circulation. 1982;66(4):777–84.
13. Thomas L, Foster E, Hoffman JI, Schiller NB. Prospective validation of an echocardiographic index for determining the severity of chronic mitral regurgitation. Am J Cardiol. 2002;90:607–12.
14. Enriquez-Sarano M, Miller FA Jr, Hayes SN, Bailey KR, Tajik AJ, Seward JB. Effective mitral regurgitant orifi ce area: clinical use and pitfalls of the proximal isovelocity surface area method. J Am Coll Cardiol. 1995;25(3):703–9.
15. Fehske W, Omran H, Manz M, Köhler J, Hagendorff A, Lüderitz B. Color-coded Doppler imaging of the vena contracta as a basis for quantification of pure mitral regurgitation. Am J Cardiol. 1994;73:268–74.
16. Hall SA, Brickner ME, Willett DL, Irani WN, Afridi I, Grayburn PA. Assessment of mitral regurgitation severity by Doppler color flow mapping of the vena contracta [see comment]. Circulation. 1997;95:636–42.
17. Helmcke F, Nanda NC, Hsiung MC, Soto B, Adey CK, Goyal RG, Gatewood RP Jr. Color Doppler assessment of mitral regurgitation with orthogonal planes. Circulation. 1987;75:175–83.
18. Utsunomiya T, Doshi R, Patel D, Nguyen D, Mehta K, Gardin JM. Regurgitant volume estimation in patients with mitral regurgitation: initial studies using color Doppler 'proximal isovelocity surface area' method. Echocardiography. 1992;9:63–70.

19. Stevenson JG. Two-dimensional color Doppler estimation of the severity of atrioventricular valve regurgitation: important effects of instrument gain setting, pulse repetition frequency, and carrier frequency. J Am Soc Echocardiogr. 1989;2(1):1–10.
20. Cape EG, Yoganathan AP, Weyman AE, Levine RA. Adjacent solid boundaries alter the size of regurgitant jets on Doppler color flow maps. J Am Coll Cardiol. 1991;17:1094–102.
21. Skjaerpe T, Hatle L. Diagnosis of tricuspid regurgitation. Sensitivity of Doppler ultrasound compared with contrast echocardiography. Eur Heart J. 1985;6:429–36.
22. Tribouilloy CM, Enriquez-Sarano M, Fett SL, Bailey KR, Seward JB, Tajik AJ. Application of the proximal flow convergence method to calculate the effective regurgitant orifice area in aortic regurgitation. J Am Coll Cardiol. 1998;32:1032–9.
23. Shiota T, Jones M, Yamada I, Heinrich RS, Ishii M, Sinclair B, Yoganathan AP, Sahn DJ. Evaluation of aortic regurgitation with digitally determined color Doppler-imaged flow convergence acceleration: a quantitative study in sheep. J Am Coll Cardiol. 1996;27:203–10.
24. Quere JP, Tribouilloy C, Enriquez-Sarano M. Vena contracta width measurement: theoretic basis and usefulness in the assessment of valvular regurgitation severity. Curr Cardiol Rep. 2003;5:110–5.
25. Tribouilloy CM, Enriquez-Sarano M, Bailey KR, Seward JB, Tajik AJ. Assessment of severity of aortic regurgitation using the width of the vena contracta: a clinical color Doppler imaging study. Circulation. 2000;102:558–64.
26. Willett DL, Hall SA, Jessen ME, Wait MA, Grayburn PA. Assessment of aortic regurgitation by transesophageal color Doppler imaging of the vena contracta: validation against an intraoperative aortic flow probe. J Am Coll Cardiol. 2001;37:1450–5.
27. Nozaki S, Mizushige K, Taminato T, Obayashi N, Matsuo H. New index for grading the severity of aortic regurgitation based on the cross-sectional area of vena contracta measured by color Doppler flow mapping. Circ J. 2003;67:243–7.
28. Smith MD, Grayburn PA, Spain MG, DeMaria AN. Observer variability in the quantitation of Doppler color flow jet areas for mitral and aortic regurgitation. J Am Coll Cardiol. 1988;11:579–84.
29. Reimold SC, Thomas JD, Lee RT. Relation between Doppler color flow variables and invasively determined jet variables in patients with aortic regurgitation. J Am Coll Cardiol. 1992;20:1143–8.
30. Taylor AL, Eichhorn EJ, Brickner ME, Eberhart RC, Grayburn PA. Aortic valve morphology: an important in vitro determinant of proximal regurgitant jet width by Doppler color flow mapping. J Am Coll Cardiol. 1990;16(2):405–12.
31. Grayburn PA, Handshoe R, Smith MD, Harrison MR, DeMaria AN. Quantitative assessment of the hemodynamic consequences of aortic regurgitation by means of continuous wave Doppler recordings. J Am Coll Cardiol. 1987;10(1):135–41.
32. Labovitz AJ, Ferrara RP, Kern MJ, Bryg RJ, Mrosek DG, Williams GA. Quantitative evaluation of aortic insufficiency by continuous wave Doppler echocardiography. J Am Coll Cardiol. 1986;8(6):1341–7.
33. Griffin BP, Flachskampf FA, Siu S, Weyman AE, Thomas JD. The effects of regurgitant orifice size, chamber compliance, and systemic vascular resistance on aortic regurgitant velocity slope and pressure half-time. Am Heart J. 1991;122(4 Pt 1):1049–56.
34. Quinones MA, Young JB, Waggoner AD, Ostojic MC, Ribeiro LG, Miller RR. Assessment of pulsed Doppler echocardiography in detection and quantification of aortic and mitral regurgitation. Br Heart J. 1980;44:612–20.
35. Rodriguez L, Thomas JD, Monterroso V, Weyman AE, Harrigan P, Mueller LN, Levine RA. Validation of the proximal flow convergence method. Calculation of orifice area in patients with mitral stenosis [see comment]. Circulation. 1993;88(3):1157–65.
36. Deng YB, Matsumoto M, Wang XF, Liu L, Takizawa S, Takekoshi N, Shimizu T, Mishima K. Estimation of mitral valve area in patients with mitral stenosis by the flow convergence region method: selection of aliasing velocity. J Am Coll Cardiol. 1994;24:683–9.
37. Rifkin RD, Harper K, Tighe D. Comparison of proximal isovelocity surface area method with pressure half-time and planimetry in evaluation of mitral stenosis. J Am Coll Cardiol. 1995;26:458–65.
38. Degertekin M, Basaran Y, Gencbay M, Yaymaci B, Dindar I, Turan F. Validation of flow convergence region method in assessing mitral valve area in the course of transthoracic and transesophageal echocardiographic studies. Am Heart J. 1998;135(2 Pt 1):207–14.