6 Left and Right Ventricular Systolic Function

The degree of ventricular systolic dysfunction is a potent predictor of clinical outcome for a wide range of cardiovascular disease, including ischemic cardiac disease, cardiomyopathies, valvular heart disease, and congenital heart disease. Echocardiographic estimates of global and regional function, quantitative ventricular volumes and ejection fractions, and Doppler echocardiographic ejection phase indices all are valuable clinical tools. Even when evaluation of ventricular systolic function is not the primary focus of the echocardiographic examination, evaluation of ventricular systolic function is a key component of every clinical study. For research applications, echocardiographic measures of left ventricular (LV) systolic function provide important baseline data on disease severity and clinical endpoints for intervention trials in patients with ventricular dysfunction.

BASIC PRINCIPLES

Cardiac Cycle

Systole is defined as the segment of the cardiac cycle from mitral valve closure to aortic valve closure (Fig. 6.1). The onset of systole is identified on the electrocardiogram as ventricular depolarization (onset of the QRS complex), with the end of systole occurring after repolarization (end of T wave). In terms of ventricular pressure curves over time, systole begins when LV pressure exceeds left atrial (LA) pressure, resulting in closure of the mitral valve. Mitral valve closure is followed by isovolumic contraction, during which the cardiac muscle depolarizes, calcium influx and myosin-actin shortening occur, and ventricular pressure rises rapidly at a constant ventricular volume

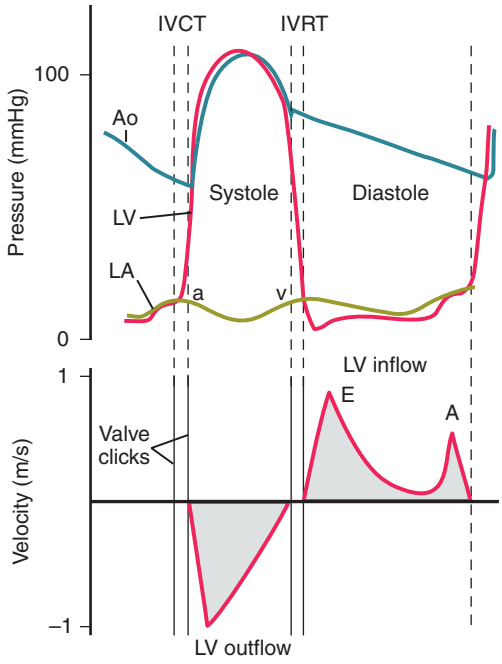


Fig. 6.1 The cardiac cycle. LV, aortic *(Ao),* and LA pressures are shown with the corresponding Doppler LV outflow and inflow-velocity curves. The isovolumic contraction time *(IVCT)* represents the time between mitral valve closure and aortic valve opening, whereas the isovolumic relaxation time *(IVRT)* represents the time between aortic valve closure and mitral valve opening.

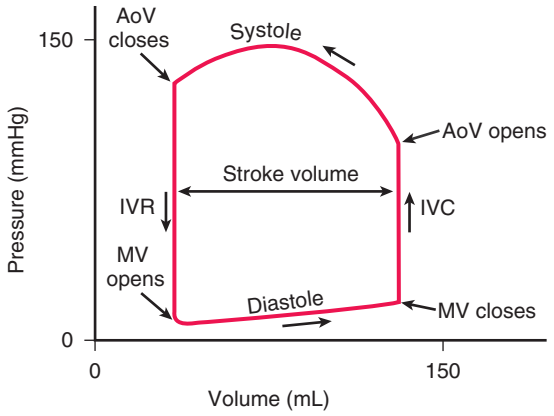


Fig. 6.2 Pressure-volume loop. LV volume is graphed on the horizontal axis, with pressure on the vertical axis. The temporal direction of pressure-volume changes is shown by the *arrows*. During diastole, volume increases with little rise in pressure. After mitral valve *(MV)* closure, isovolumic contraction *(IVC)* results in a rapid rise in pressure with no change in volume. At the onset of ejection, the aortic valve *(AoV)* opens with a rapid decrease in LV volume during systole. Aortic valve closure is followed by isovolumic relaxation *(IVR)*.

(although shape changes do occur). When ventricular pressure exceeds aortic pressure, the aortic valve opens. During ejection (aortic valve opening to closing), LV volume falls rapidly as blood flows from the LV to the aorta. LV pressure exceeds aortic pressure for approximately the first half of systole, corresponding to a rapid acceleration of blood flow and a small pressure difference from the ventricle to the aorta. In the normal heart, pressure crossover occurs in mid-systole, so during the second half of systole, aortic pressure exceeds LV pressure, thus resulting in continued forward blood flow but at progressively slower velocities (deceleration). Aortic valve closure occurs at the dicrotic notch of the aortic pressure tracing, immediately following end-ejection. In sum, systole includes isovolumic contraction and ventricular ejection (acceleration and deceleration phases). Ventricular volume ranges from a maximum at end-diastole (or onset of systole) to a minimum at end-systole.

Physiology of Systolic Function

During systole, ventricular myocardial fibers contract circumferentially and longitudinally, resulting in myocardial wall thickening and inward motion of the endocardium. The simultaneous decrease in ventricular size and increase in pressure result in ejection of a volume of blood (stroke volume) from the ventricle. Stroke volume reflects the *pump performance* of the heart. The decrease in chamber volume relative to end-diastolic volume, or ejection fraction, reflects overall *ventricular function*. Ventricular function and pump performance depend on:

- Contractility (the basic ability of the myocardium to contract)
- Preload (initial ventricular volume or pressure)
- Afterload (aortic resistance or end-systolic wall stress)
- Ventricular geometry

Contractility is the intrinsic ability of the myocardium to contract, independent of loading conditions or geometry. Evaluation of contractility itself thus requires measurement of ventricular ejection performance under different loading conditions. Experimentally, contractility often is described by the slope of the end-systolic pressure-volume relationship (E_{max}). To derive this value, LV pressure is graphed on the vertical axis, with volume (not time) on the horizontal axis (Fig. 6.2). This pressure-volume "loop" then represents a single cardiac cycle, with different pressure-volume loops for the same ventricle representing different loading conditions (e.g., increasing or decreasing ventricular end-diastolic volume or changing afterload). E_{max} is the slope of the line that intersects the end-systolic pressure-volume point for each curve. A decrease in contractility results in a decrease in stroke volume and larger LV volumes (Fig. 6.3). Contractility itself can be affected by several physiologic parameters including heart rate, coupling interval, and metabolic factors, in addition to disease processes and pharmacologic agents.

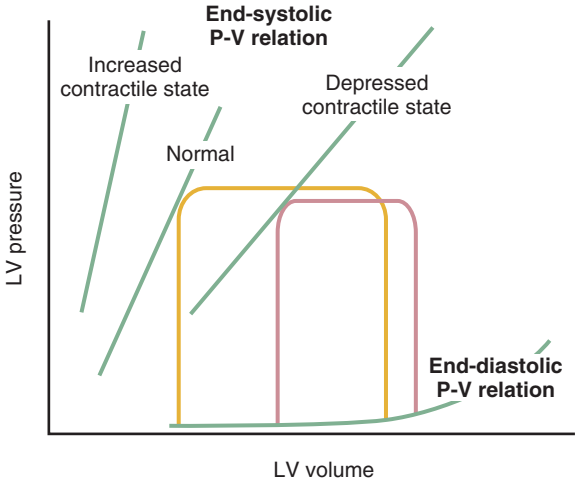


Fig. 6.3 Effect of changes in contractility on left ventricular pressure-volume loops. A normal ventricle is shown in yellow and an acute decrease in contractile state in pink. The slope of the line intersecting the end-systolic pressure-volume (P-V) points at different loading conditions, shown as a green line for each P-V loop, is a measure of contractility known as elastance (E_{max}), which is insensitive to changes in loading conditions. With decreased contractility, the P-V loop is displaced to the right, and the end-systolic P-V line shifts downward and to the right. The effect of an acute increase in contractile state is illustrated by the line on the left; the slope of the end-systolic P-V line is increased (the corresponding loop is not shown). *(From Aurigemma GP, Gaasch WH, Villegas B, et al: Noninvasive assessment of left ventricular mass, chamber volume, and contractile function.* Curr Probl Cardiol *20:418, 1995.)*

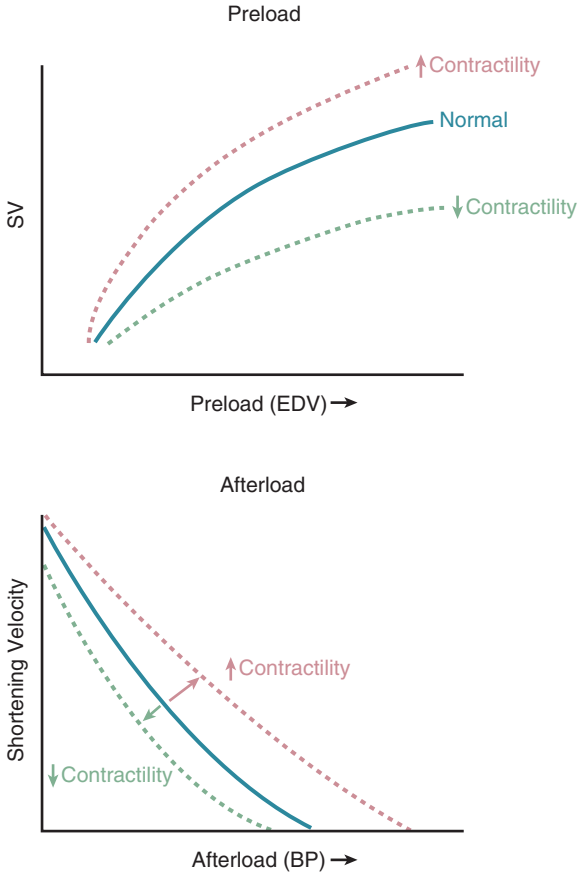
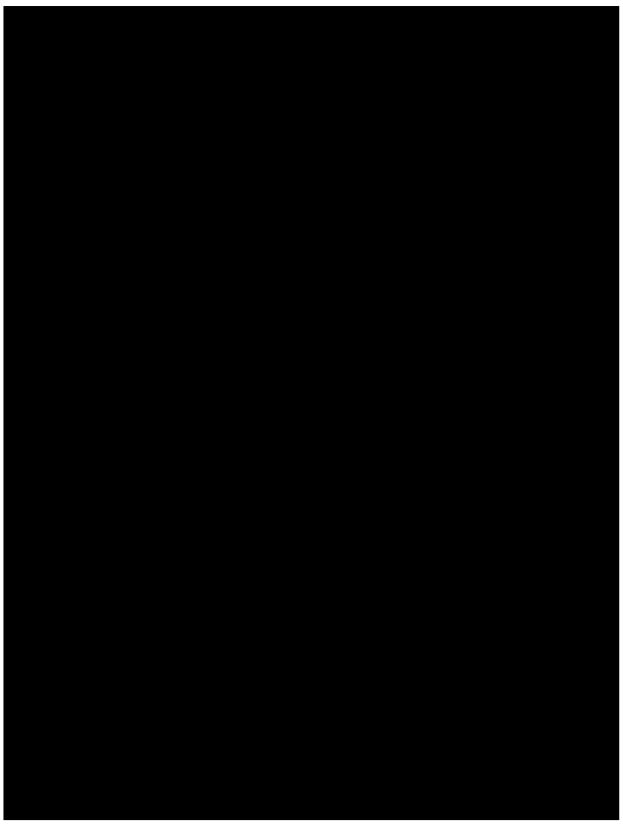


Fig. 6.4 Preload and afterload. *(Top)* The relationship between preload, often defined by end-diastolic volume *(EDV)*, and stroke volume *(SV)* is shown for a normal *(blue line)* LV. With increased contractility, one sees a greater increase in SV for an increase in EDV *(red line)*; with decreased contractility, one sees a smaller increase in SV for an increase in EDV *(green dashed line). (Bottom)* The inverse relationship between afterload, approximated by blood pressure *(BP)* or systemic vascular resistance, and LV myocardial shortening velocity is shown for a normal ventricle *(blue line).* With increased contractility, shortening velocity (and stroke volume) can be maintained at higher afterloads *(red line)*; with decreased contractility, shortening velocity is lower for any given afterload *(green dashed line).*

The effect of *preload* on ventricular ejection performance is summarized by the Frank-Starling curve showing ventricular end-diastolic volume (or pressure) on the horizontal axis and stroke volume on the vertical axis (Fig. 6.4). For a given degree of contractility, a curvilinear relationship exists between these variables such that increasing end-diastolic volume results in a greater stroke volume. An increase in contractility results in a greater increase in stroke volume for a given increase in preload; a decrease in contractility has the opposite effect.

Afterload, defined by resistance or impedance, has an inverse relationship with myocardial fiber shortening such that increasing vascular resistance results in a decreased stroke volume (see Fig. 6.4). An increase in contractility allows maintenance of a normal stroke volume with a higher afterload. With a decrease in contractility, even slight increases in afterload further decrease myocardial fiber shortening and stroke volume.

Measurement of LV systolic function independent of loading conditions is difficult using echocardiographic or other clinical approaches. It rarely is possible to construct pressure-volume loops under different loading conditions because of the problem of measuring instantaneous LV volumes and the potential risk of altering loading conditions in ill patients. Thus clinical evaluation of ventricular function has focused on measurements of cardiac output, ejection fraction, and end-systolic dimension or volume, even though the load dependence of these measures is a clearly acknowledged limitation. Strain and strain rate measurements offer another approach to evaluation of ventricular function and are becoming more widely used.



DOPPLER EVALUATION OF LEFT VENTRICULAR SYSTOLIC FUNCTION

Stroke Volume Calculation

Doppler echocardiographic evaluation of LV systolic function usually is based on calculation of stroke volume and cardiac output (see Appendix B, Table B.3).

ECHO MATH: Doppler Stroke Volume and Cardiac Output

Using Doppler and 2D echo data, stroke volume (SV in cm^3 or mL) is calculated as cross-sectional area (CSA in cm^2) of flow times the velocity-time integral (VTI in cm) of flow through that region:

$$SV = CSA \times VTI \quad \text{(Eq. 6.11)}$$

and

$$CO = SV \times HR \quad \text{(Eq. 6.12)}$$

Where CO is cardiac output and HR is heart rate.

For example, if LV outflow tract diameter is 2.3 cm, LV outflow-velocity time integral is 11 cm, and heart rate is 88 bpm:

Cardiac output (CO) is calculated using the LVOT diameter to calculate the circular cross-sectional areas of flow:

$$CSA_{LVOT} = \pi(LVOT_D/2)^2 = 3.14(2.3/2)^2 = 4.2\ cm^2$$

Stroke volume across the aortic valve (cm^3 = mL), then, is:

$$SV_{LVOT} = (CSA_{LVOT} \times VTI_{LVOT}) = 4.2\ cm^2 \times 11\ cm = 46\ cm^3$$

Cardiac output is:

$$CO = SV \times HR = 46\ mL \times 88\ beats/min = 4048\ mL/min \text{ or } 4.05\ L/min$$

Cardiac index (CI) normalizes flow to body surface area (BSA):

$$CI = CO/BSA = 4.05\ L/min/1.8\ m^2 = 2.25\ L/min/m^2$$

This indicates a low cardiac output and index (normal >2.5 $L/min/m^2$)

Conceptually, the LV ejects a volume of blood into the cylindrical aorta on each beat (Fig. 6.13). The base of this cylinder is the systolic cross-sectional area of the outflow tract, whereas its height is the distance the average blood cell traveled during ejection for that beat. This distance is expressed as the integral of the Doppler systolic velocity-time curve because velocity is the first derivative of distance. Alternatively, this distance also can be thought of as mean velocity (cm/s) multiplied by ejection duration (seconds). Again, because the volume of a cylinder is base times height, stroke volume is cross-sectional area multiplied by the velocity-time integral.

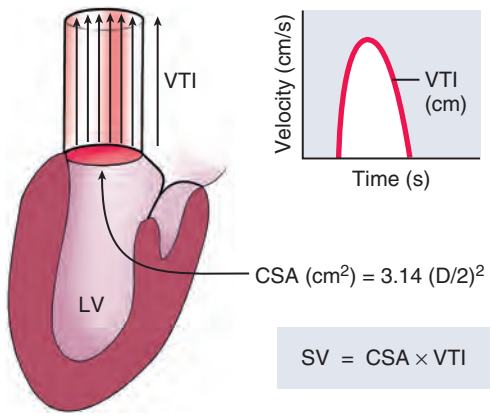


Fig. 6.13 Doppler stroke volume concept. The cross-sectional area *(CSA)* of flow is calculated as a circle based on a 2D echo diameter *(D)* measurement. The length of the cylinder of blood ejected through this cross-sectional area on a single beat is the velocity-time integral *(VTI)* of the Doppler curve. Stroke volume *(SV)* then is calculated as CSA × VTI.

This approach to stroke volume calculation depends on several basic assumptions:

- Accurate cross-sectional flow area measurement
- Laminar flow with spatially "flat" flow-velocity profile
- Parallel intercept angle between Doppler beam and direction of blood flow
- Velocity and diameter measurements are made at the same anatomic site

First, the cross-sectional area must be measured accurately. Typically, diameter is measured and 2D area calculated as $\pi(D/2)^2$ based on the assumption of a circular geometry. Deviations from a circular geometry or changes in cross-sectional area during the flow period will result in inaccuracies unless appropriate corrections are included in the calculations. Small errors in 2D diameter measurements become large errors in cross-sectional area calculations because radius (half the diameter) is squared in the calculations. Using a transducer orientation and instrument settings that maximize image quality, performing measurements based on axial (rather than lateral) resolution, performing diameter measurements in two orthogonal planes (when possible), and averaging several beats can help minimize this source of error.

Second, the pattern of flow is assumed to be laminar, and the spatial flow profile across the flowstream is assumed to be relatively flat. These assumptions ensure that the velocity curve represents the spatial (and temporal) average flow in that region. The validity of the assumption of laminar flow in the great vessels and across normal cardiac valves is demonstrated by the narrow band of velocities and smooth spectral signal seen on pulsed Doppler echo recordings. A flat flow profile also is a reasonable assumption at the inlet to the great vessels and across the valve planes because of the effects of geometric convergence and acceleration. A flat flow-velocity profile can be confirmed by moving the sample volume across the flowstream in two orthogonal views to demonstrate uniform velocities at the center and the edges of the flowstream.

Third, the Doppler signal is assumed to have been recorded at a parallel intercept angle to flow, resulting in an accurate velocity measurement (based on a cos $\theta = 1$ in the Doppler equation). In practical terms, the sonographer aligns the Doppler beam in the presumed direction of flow and then carefully moves the ultrasound beam across the image plane and in the elevational plane to obtain the highest-velocity signal, thus indicating the most parallel alignment with flow. Note that the optimal window for Doppler interrogation is when the ultrasound beam and flowstream are parallel, whereas the optimal window for diameter measurement is when the ultrasound beam and tissue-blood interfaces are perpendicular.

Fourth, it is crucial that the diameter and velocity measurements be made at the same anatomic site because the cross-sectional area and flow-velocity curves must be temporally and spatially congruent for accurate volume flow rate calculations. As the cross-sectional area of flow narrows or expands, flow velocity will increase or decrease correspondingly, so that conjoining information from two different anatomic sites will result in erroneous stroke volume data. Similarly, dynamic changes in stroke volume occur with changes in heart rate, loading conditions, exercise, and so on, so that measurements made at disparate times cannot be combined. In clinical practice, diameter and velocity recordings are made in close sequence and are repeated if any question of an interval physiologic change exists.

Sites for Stroke Volume Measurement

Stroke volume can be measured by this approach at any intracardiac site where both cross-sectional area and the flow-velocity integral can be recorded, given the assumptions of laminar flow and a flat flow profile.

Left Ventricular Outflow

The standard site for stroke volume measurement is the LV outflow tract at the level of the aortic annulus just proximal to the valve leaflets. The LV outflow tract offers several advantages: (1) the spatial flow profile is relatively flat because of tapering geometry and flow acceleration, (2) the needed data can be recorded in nearly all patients, and (3) flow remains laminar proximal to a stenosis (allowing transaortic stroke volume calculations in patients with aortic valve disease). LV outflow tract diameter is measured in a parasternal long-axis view parallel and immediately adjacent to the aortic valve, in mid-systole, from the white-black edge of the septal endocardium to the black-white edge of the anterior mitral leaflet (Fig. 6.14). Pulsed Doppler is used from an apical approach to record the velocity curve, by using the closing click of the aortic valve to ensure that the sample volume is located at the annulus (the same site as the diameter measurement). The small region of flow convergence proximal to the narrowed aortic valve is avoided by moving the sample volume slightly apically until a narrow spectral width is seen at the velocity peak.

On TEE imaging, outflow tract diameter is measured in a long-axis view with improved accuracy because of the higher-resolution images on TEE. LV outflow velocity sometimes may be recorded from a transgastric apical view or, starting from a transgastric short-axis view, rotating the image plane 90° to the two-chamber view and then turning the transducer slightly medially to visualize the LV outflow tract. However, it is difficult to ensure a parallel intercept angle between the ultrasound beam and LV outflow, so underestimation of stroke volume is likely.

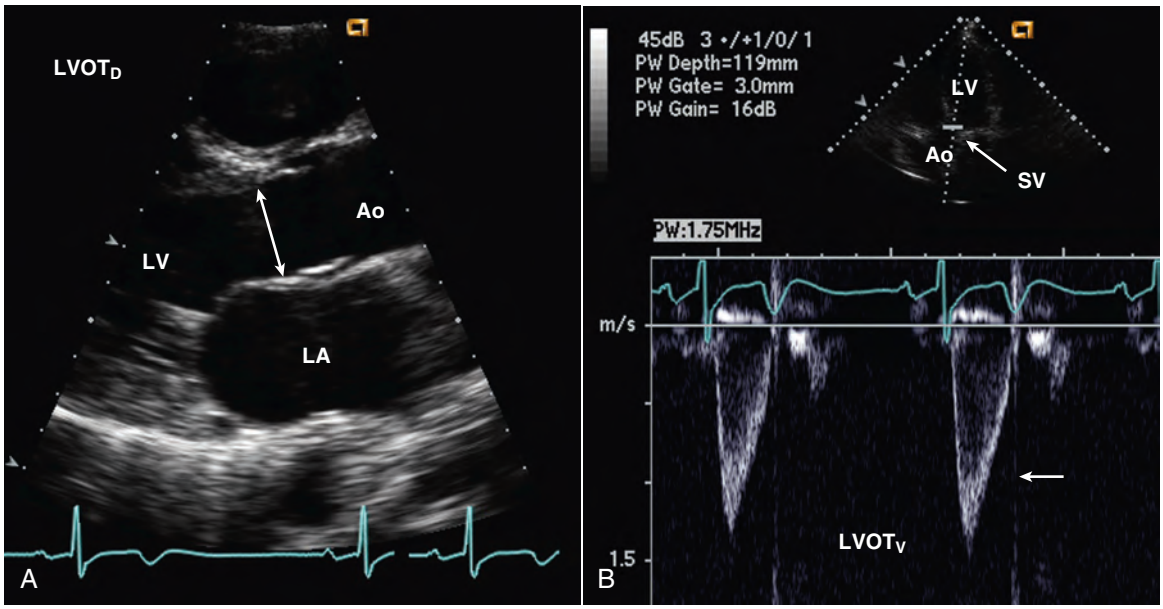


Fig. 6.14 Doppler left ventricular outflow stroke volume calculation. (A) LV outflow tract diameter *(LVOT$_D$)* measurement from a parasternal long-axis view and (B) the pulsed Doppler recording of LV outflow just proximal to the aortic valve *(Ao)* from an apical approach for stroke volume calculation. The aortic valve closing click *(arrow)* on the outflow tract velocity *(LVOT$_V$)* recording ensures that the sample volume *(SV)* location is immediately adjacent to the valve, corresponding with the site of outflow tract diameter measurement.

LV stroke volume also can be measured in the ascending aorta with diameter measured from a parasternal long-axis view and the flow-velocity curve recorded from either an apical or a suprasternal notch window. If continuous-wave (CW) Doppler ultrasound is used, the highest velocities along the path of the beam will be recorded, so the narrowest segment of the aorta (the sinotubular junction) is used for diameter measurements. If pulsed Doppler is used, the aortic diameter measurement should correspond to the Doppler sample volume location. Note that if aortic valve disease is present, stroke volume measurement in the ascending aorta will be inaccurate because of nonlaminar flow distal to the valve.

Mitral Valve

Transmitral stroke volume calculations assume that the mitral annulus is the limiting cross-sectional flow area, with the leaflets moving passively in response to the flowstream. Transmitral stroke volume is calculated as the product of the cross-sectional annulus area and the velocity-time integral of flow recorded at the mitral annulus level. On TEE imaging, transmitral flow rate is measured in the four-chamber view with the pulsed Doppler sample volume placed at the annulus level and the diameter measured from the 2D image (Fig. 6.15). Although the mitral annulus is most accurately described as a curved ellipse with the major (more apical) axis seen in the four-chamber view and the minor (more basal) axis seen in the long-axis view, for most clinical applications, the mitral annulus is assumed to be circular. Mitral annulus diameter can be measured in a parasternal long-axis view, which has the advantage of using axial resolution (which improves accuracy) but the disadvantage of ambiguity in the correct measurement location because the site of Doppler recording must be estimated. Alternatively, diameter can be measured in the apical four-chamber view, which has as the advantage that diameter can be measured on the same image that displays the sample volume (ensuring a correct measurement site) but the disadvantage of lateral resolution, which limits the accuracy of the measurement.

Right Heart

Stroke volume can be calculated by analogous methods in the pulmonary artery or across the tricuspid valve (Fig. 6.16). In adult patients, use of the pulmonary artery site on TTE imaging often is limited by poor image quality, which results in unobtainable or inaccurate pulmonary artery diameter measurements. However, from a TEE approach, pulmonary artery flow and diameter often can be measured using a very high transducer position looking from the pulmonary artery bifurcation toward the pulmonic valve.

Differences in Transvalvular Volume Flow Rates

In a normal heart, stroke volume across each of the four valves is equal, and measurement at more than

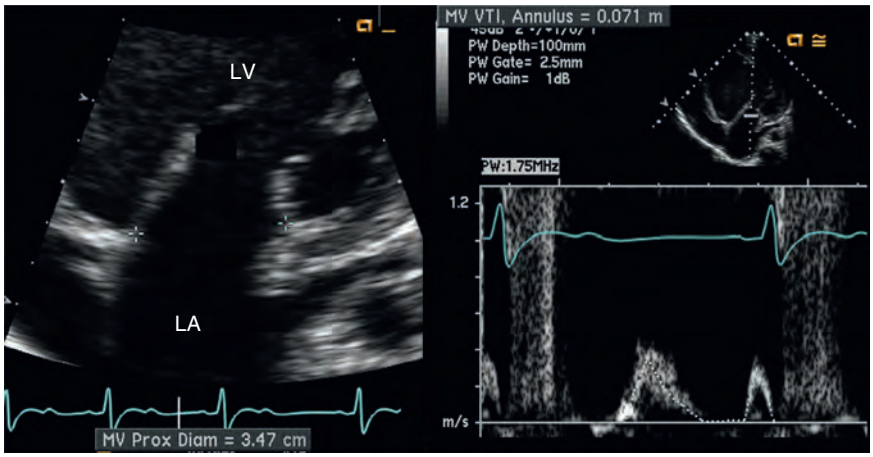


Fig. 6.15 Doppler transmitral stroke volume calculation. The mitral valve *(MV)* annulus area in diastole is multiplied by the velocity-time integral *(VTI)* of flow at the annulus. In this example, annulus diameter *(left)* is 3.5 cm, so the circular cross-sectional area is 9.6 cm^2. The velocity-time integral of transmitral flow is 7.7 cm for a stroke volume of 74 mL or a cardiac output of 4.4 L/min at a heart rate of 60 bpm.

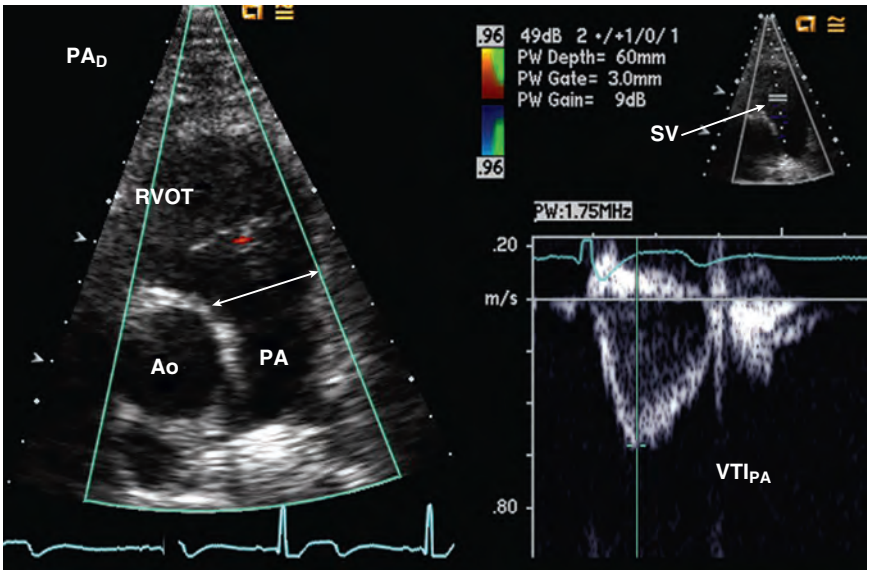


Fig. 6.16 Doppler stroke volume in the pulmonary artery. Diameter is measured in a parasternal RV outflow view *(RVOT; left)* and pulsed Doppler pulmonary artery *(PA)* flow recorded from a parasternal approach *(right)* for transpulmonic stroke volume *(SV)* calculation. PA_D, pulmonary artery diameter; VTI_{PA}, pulmonary artery velocity-time integral.

one site only serves as an internal accuracy check. However, in the presence of valvular regurgitation or an intracardiac shunt, calculation of stroke volume at two intracardiac sites allows quantitation of the degree of regurgitation or pulmonic-to-systemic shunt ratio, as detailed in Chapters 12 and 17.

Other Doppler Measures of Systolic Function

Ejection Acceleration Times

In addition to stroke volume calculations, the shape of the Doppler ejection curve provides information about ventricular function. When systolic function is normal, the isovolumic contraction period is short, and the rate of pressure rise in early systole is rapid. These features are reflected in the Doppler velocity curve, which shows a short isovolumic contraction time, a rapid acceleration of blood in early systole, and a short time interval from the onset of flow to maximum velocity. With impaired LV systolic function, the isovolumic contraction time (also known as the pre-ejection period) becomes progressively longer, the rate of acceleration diminishes, and the time to maximum velocity increases, with all these changes mirrored in the Doppler velocity curve. In addition to measuring these variables at rest, some centers have found evaluation of aortic ejection