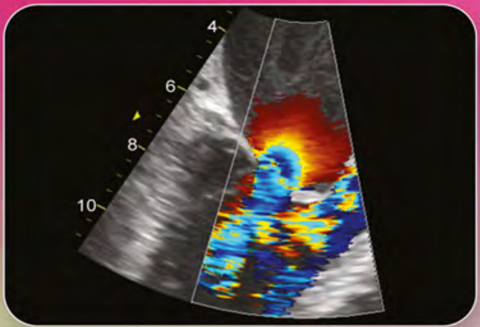
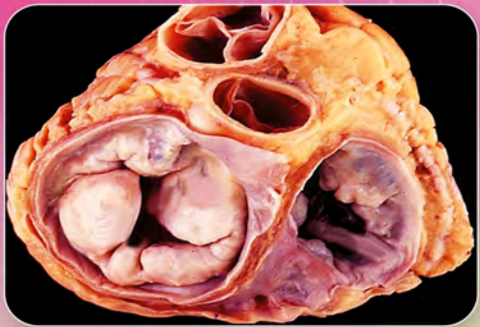
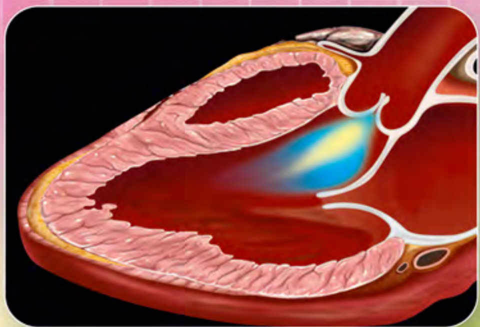
A Practical Approach to Clinical Echocardiography







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Foreword
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CHAPTER 2

Hemodynamic Evaluation by Echo-Doppler Techniques

❐ INTRODUCTION

Hemodynamics can be defined as the physical factors that govern blood flow. These are the same physical factors that govern the flow of any fluid, and are based on a fundamental law of physics. Pressure, flow and resistance are all fundamental elements of fluid mechanics. The relationship between these parameters clearly defines the behavior of the blood in the heart and vessels of the human body. Disturbance in flow, intracardiac pressure and resistance all provide useful information in health and disease, which can be used to diagnose severity of valvular stenosis and regurgitation, shunt calculations, assess the integrity of pulmonary circulation, monitor therapy and long-term follow-up of asymptomatic patients with altered pressure-flow dynamics due to structure heart disease. The valves ensure that blood flows in a single pathway through the heart by opening and closing in a particular time sequence during the cardiac cycle and this mechanism may not be true even in normal conditions. How much of this "valve mechanism" is abnormal can be studied by Doppler hemodynamics. Thus, hemodynamics is an important part of cardiovascular physiology dealing with the forces the pump (the heart) has to develop to circulate blood through the cardiovascular system in normodynamics and otherwise.

Echocardiography has replaced invasive cardiac catheterization for various hemodynamic assessments: cardiac output, valvular pressure gradients, dynamic LV outflow tract gradient, right heart pressures, constrictive pericarditis and vascular resistance.

❐ HEMODYNAMIC EVALUATION

1. *Hemodynamics*: Is an important part of cardiovascular physiology dealing with the forces the pump (the heart) has to develop to circulate blood through the cardiovascular system. Both normotension and normodynamic state (blood flow within normal range) must be part of the therapeutic goal.[1]

2. *Flow*: Blood in the circulatory system moves from one point to another driven by a pressure difference at the two points. It can be expressed as total flow volume in one cardiac cycle either in systole or diastole or flow rate (flow/time). Speed with which it moves and the direction are expressed as velocity. Velocity of flow can be obtained by using Doppler echocardiography.[2] Flow is a product of velocity and area of flow (Fig. 2.1).

3. *Velocity*: Velocity basically is the rate of change of the position of an object with time. For blood, it denotes the speed and direction of flow (cm/s).

4. *Acceleration*: Rate of change in velocity is called acceleration (cm/s^2).

5. *Force*: is a product of mass and acceleration. As the mass is constant, force and acceleration can be used interchangeably.

6. *Pressure*: Force per unit area is called the pressure.

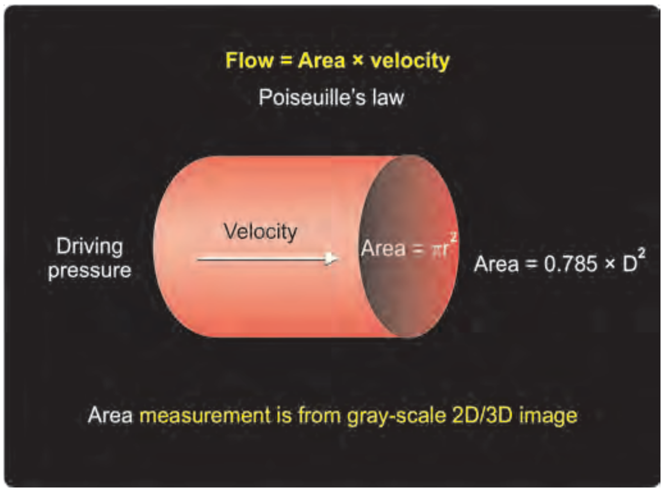

Fig. 2.1: Flow through a cylindrical object is expressed by velocity × area. Area is obtained from the two- or three-dimensional gray-scale image. For a spherical structure, area = 0.785 × (diameter)².

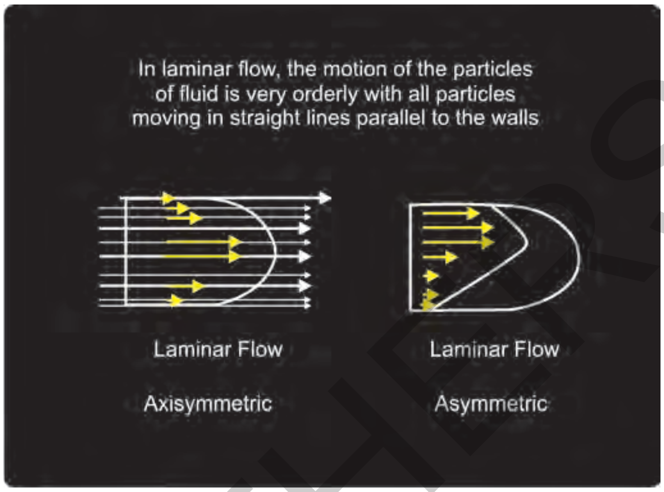

Fig. 2.2: Laminar flow through a cylinder can be either axisymmetric or asymmetric.

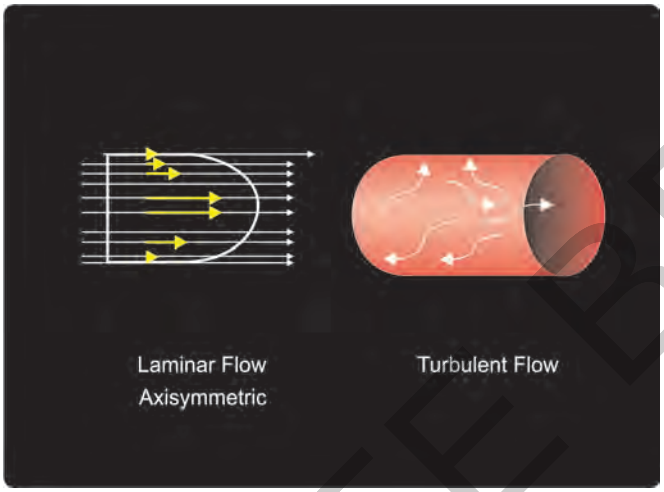

Fig. 2.3: Laminar flow versus turbulent flow.

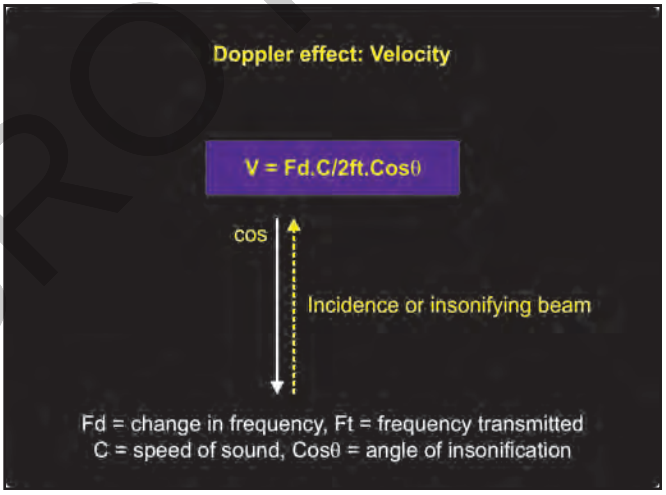

Fig. 2.4: Doppler equation.

Types of Flow

Flow can be laminar or turbulent. Laminar flow runs in parallel streamlines, while the turbulent flow is chaotic. It is difficult to apply laws of physics on turbulent flow. Estimation of flow using various laws of physics assumes that the flow is in streamline (Figs 2.2 and 2.3).

Turbulence and its extent are defined by the Reynolds number.

❑ DOPPLER PRINCIPLE FOR ESTIMATING VELOCITY

If speed of sound through a medium is known or presumed, velocity of flow in it can be determined by the change in frequency of the moving objects (red blood cells) when compared to the transmitting frequency (Fig. 2.4).

In practice, speed of sound is presumed to be 1,540 m/s, and angle of insonification is kept as close to zero as possible so that no angle calculation or correction is required. Angle correction has been given up except while studying vascular flow. If frequency shift is positive (received frequency > transmitted frequency), velocity is positive toward the transducer and vice versa.

Types of Doppler Echo

Types of Doppler are:
1. Pulsed wave Doppler
2. Continuous wave Doppler
3. Color Doppler
4. Tissue Doppler
5. Power Doppler

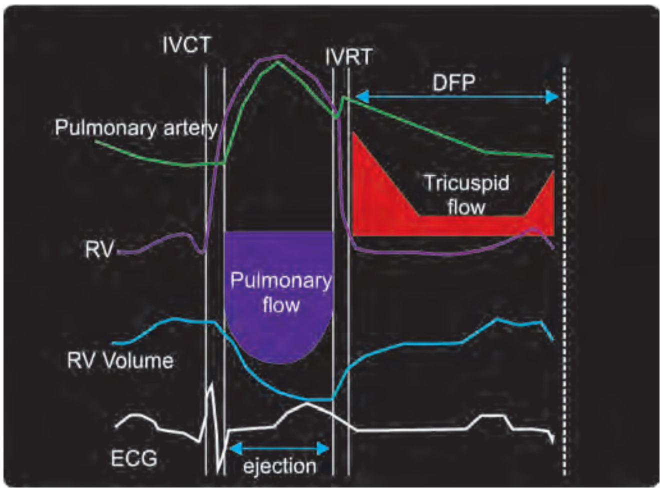

Fig. 2.5: The right heart circulation has low velocity inflow across the tricuspid valve in diastole and low velocity outflow across the pulmonary valve in systole (approximately 100 cm/s).

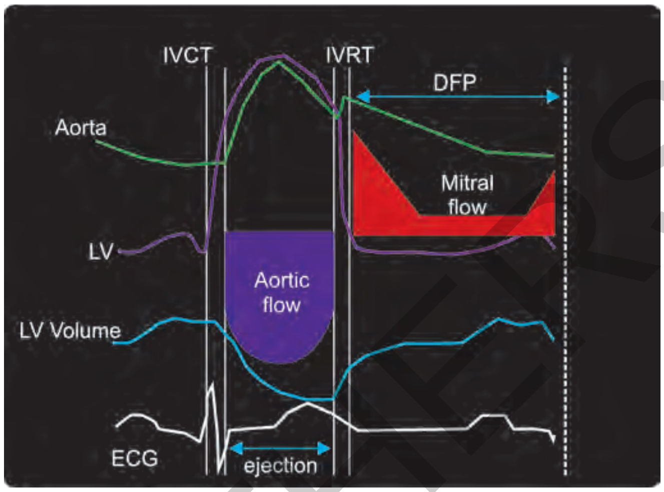

Fig. 2.6: The left heart circulation shows low-velocity biphasic inflow across the mitral valve (< 100 cm/s) and low-velocity outflow across the aortic valve in systole (< 150 cm/s).

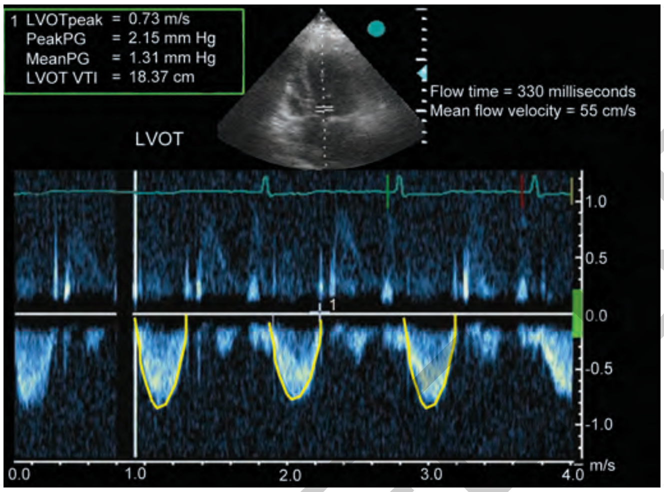

Fig. 2.7: Left ventricular outflow tract peak velocity (apical view) by pulsed wave Doppler and calculation of LVOT–velocity–time integral (VTI). LVOT–VTI (stroke distance), herein, is 18.37 cm. Stroke volume is obtained when LVOT–VTI is multiplied by the cross-sectional area of the outflow tract.

PW uses a single crystal that sends and receives sound beam signals.

- The crystal emits a short burst of ultrasound at a certain frequency [pulse repetition frequency (PRF)].
- After a certain interval determined by the depth of interest, it samples the backscatter signals.
- The maximal frequency shift that can be determined is half the PRF or Nyquist limit.
- If the velocity exceeds the Nyquist limit, aliasing occurs—Doppler spectrum gets cut off at the Nyquist frequency and the rest is recorded in the opposite direction.

CW uses two piezoelectric crystals—one continuously transmits and another continuously receives.

- Useful for high velocities, not limited by the PRF or Nyquist phenomenon
- To get high velocities, one can use the nonimaging transducer.

The normally functioning valves ensure that blood flows in a single pathway through the heart by opening and closing in a particular time sequence during the cardiac cycle and adequate flow is maintained at low-flow velocity and driving pressure because valves offer very little resistance (Figs 2.5 and 2.6). This type of flow is obtained by PW Doppler.

Velocity–Time Integral

For estimation of mean flow, velocity needs to be multiplied by time to obtain velocity–time integral (VTI). VTI is also known as stroke distance.[2] Stroke distance when multiplied by cross-sectional area (CSA), provides volume (Fig. 2.7).

VTI is used to estimate:

- Stroke volume (SV) and cardiac output
- Regurgitant volumes and regurgitant fractions
- Qp:Qs (ratio of left-to-right shunt or pulmonary blood flow to systemic flow ratio).

Qp:Qs is calculated after obtaining the right ventricular stroke volume [right ventricular outflow tract

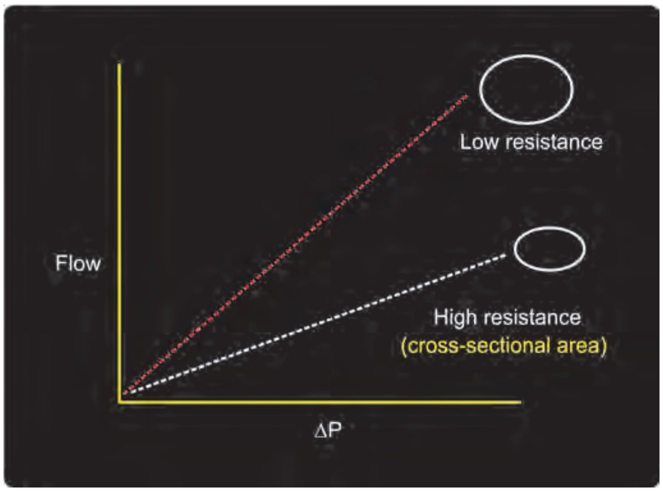

Fig. 2.8: Relationship between flow and driving pressure and resistance.

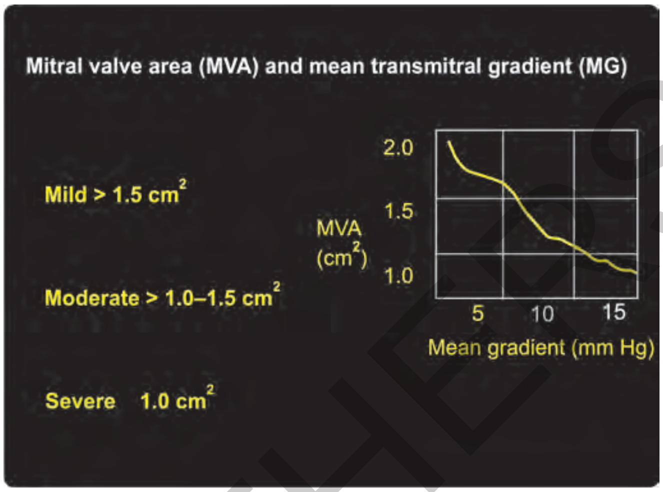

Fig. 2.9: An approximate relationship between mitral valve area (MVA) and transmitral mean diastolic gradient in subjects with normal flow.

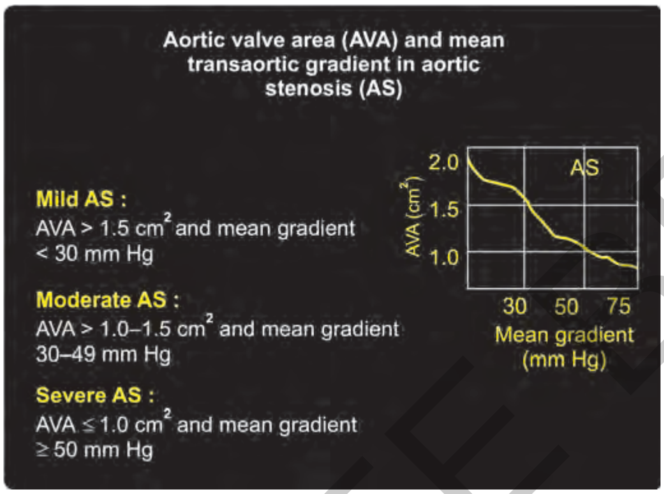

Fig. 2.10: An approximate relationship between aortic valve area and mean transaortic gradient in subjects with normal flow.

(RVOT)-VTI × RVOT area], the left ventricular stroke volume [left ventricular outflow tract (LVOT)-VTI × LVOT area] and then getting their ratio. In subjects with no intracardiac shunts or valvular regurgitation, Qp = Qs. Absolute left-to-right shunt can be calculated by multiplying stroke volume difference by heart rate when the right ventricular stroke volume is definitely more than the left ventricular (LV) stroke volume.

Cardiac output (CO) is calculated as: CO = SV × heart rate (L/min).

Normal stroke volume is 45 ± 5 mL/M^2. Low flow is labeled when stroke volume is < 35 mL/M^2.

❒ OHM'S LAW OF FLUIDS (THE HAGEN–POISEUILLE EQUATION)

The Ohm's law when applied to fluids suggests that

Driving pressure (ΔP) = Flow (Q) × resistance (R)

In relating Ohm's law to fluid flow,[3] the voltage difference is the pressure difference (ΔP; sometimes called driving pressure, perfusion pressure or pressure gradient). For the flow of blood in a chamber or across it, the ΔP is the pressure difference between any two points along a given length. There is a linear and proportionate relationship between flow and driving pressure. This linear relationship, however, is not followed when pathological conditions lead to turbulent flow because turbulence decreases the flow at any given driving pressure. Because flow and resistance are reciprocally related, an increase in resistance decreases flow at any given ΔP (Fig. 2.8). Also, at any given flow along a blood vessel or across a heart valve, an increase in resistance increases the ΔP.

The resistance (R) is the resistance to flow that is related in large part to the size of the valve opening or any other orifice.

Driving pressure or ΔP can be obtained in the heart using the modified Bernoulli's equation.[4] Resistance is the ratio of pressure difference/flow (in Wood units or dynes.cm.s^{-5}).

There is an approximate relationship between narrowed valve area and mean gradients (Figs 2.9 and 2.10).

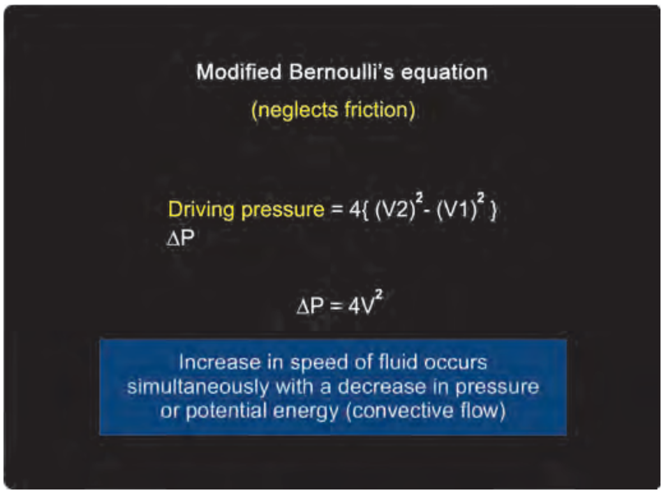

Fig. 2.11: Using Doppler principle, ΔP can be estimated by a difference in velocities at two points. If the proximal point velocity is neglected (being very low due to low resistance), the equation is simpl fied to $\Delta P = 4V^2$ wherein V represents the maximum velocity obtained at the distal point.

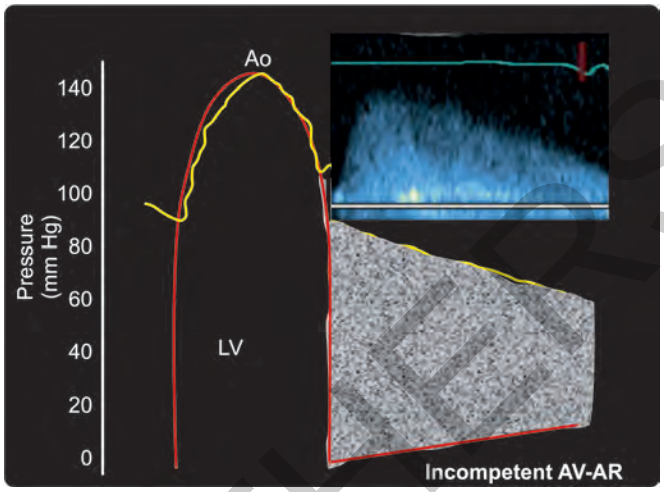

Fig. 2.12: Driving pressure during diastole in an incompetent aortic valve. The right upper corner shows continuous wave Doppler tracing, which is a replica of ΔP and flow during diastole shown graphically defining aortic regurgitation. At the onset of regurgitant flow, ΔP is 80 mm Hg, which decreases to 40 mm Hg at the end of flow.

❒ THE BERNOULLI'S EQUATION

Doppler echocardiography takes advantage of the acceleration of flow across a restrictive orifice, and the relationship defined by the Bernoulli's equation between velocity and pressure, to assess gradients.

Increase in speed of fluid occurs simultaneously with a decrease in pressure or potential energy (convective flow), provided there are minimal frictional forces. Within a fluid flowing horizontally, the highest speed occurs where pressure is the lowest, and the lowest speed occurs where pressure is the highest (Fig. 2.11).

It is an approximate relation between pressure, velocity and elevation, and is valid in regions of steady, incompressible flow where net frictional forces are negligible. The Bernoulli's equation can not be applied to normal transvalvular flows because accelerative flow is more than the convective flow.

The Bernoulli's principle is derived from the principle of conservation of energy. This states that, in a steady flow, the sum of all forms of mechanical energy in a fluid along a streamline is the same at all points on that streamline. This requires that the sum of kinetic energy and potential energy remain constant. Thus, an increase in the speed of the fluid occurs proportionately with an increase in both its dynamic pressure and kinetic energy, and a decrease in its static pressure and potential energy (Fig. 2.12).

ΔP

- Doppler echo measures blood-flow velocities, which can be converted to pressure gradients by the Bernoulli's equation.
- In most clinical situations, flow acceleration and viscous friction components of the equation can be ignored.
- Flow velocity proximal to a fixed orifice (v_1) is much lower than the peak; so this can be ignored as well.
- Pressure gradient (ΔP) across a fixed orifice can then be calculated with the simplified Bernoulli's equation. *Simplified Bernoulli's equation*: $\Delta P = 4v^2$

Caveats for the Bernoulli's Equation

- Proximal velocity may not be negligible.
- There may be multiple sites across the pathway of flow, where relationship of kinetic energy to potential energy changes.
- Frictional forces may not be negligible.
- Flow may not always be in streamlines.
- Accelerative forces may be significant depending upon the shape of the orifice through which the flow exits.

❒ CONTINUITY EQUATION

One of the fundamental principles used in the analysis of uniform flow is known as the Continuity of Flow.[5]

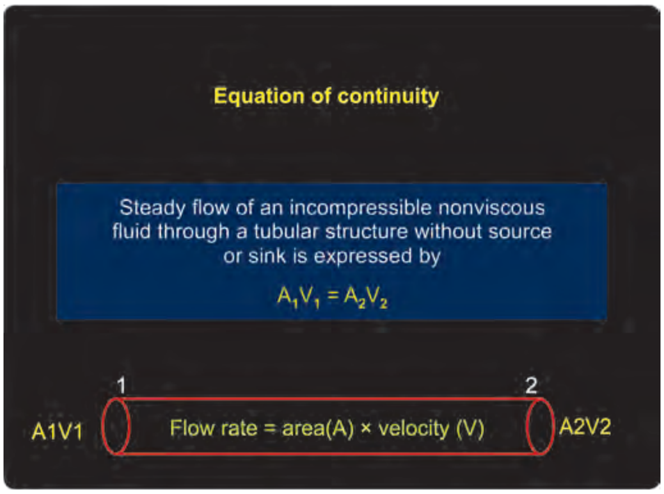

Fig. 2.13: Flow across a pathway is the product of velocity and area. Unless area changes (changing the velocity), the flow at two points is constant.

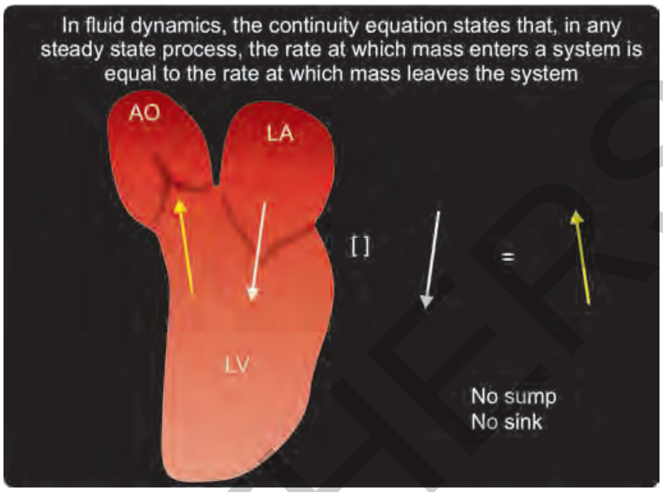

Fig. 2.14: Left heart: The rate at which mass enters the system (mitral inflow) is equal to the rate at which mass leaves the system (outflow). This is the equation of continuity. As mass is volume × density and the density of blood is constant, the equation can be applied to the volume entering and leaving the system. provided there is no sink or sump in the system.

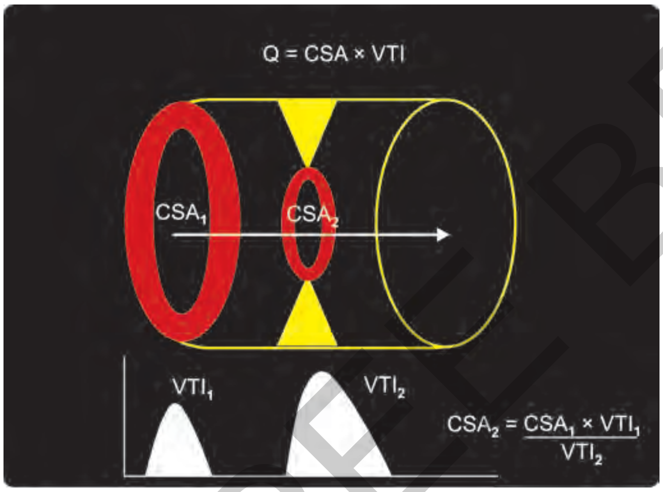

Fig. 2.15: Continuity equation to estimate cross-sectional area (CSA) at point 2. (Q: Flow volume; VTI: Velocity–time integral).

This principle is derived from the fact that mass is always conserved in fluid systems regardless of the pathway complexity or direction of flow.

If steady flow exists in a channel and the principle of conservation of mass is applied to the system, there exists a continuity of flow, defined as follows: The mean velocities at all cross sections having equal areas are then equal, and if the areas are not equal, the velocities are inversely proportional to the areas of the respective cross sections (Fig. 2.13).

Thus, if the flow is constant in a reach of channel, the product of the area and velocity will be the same for any two cross sections within that reach. This is expressed in the continuity equation (Fig. 2.14).

If the velocities (or velocity–time integrals) at two points are estimated by the Doppler equation and area of one point is known, the area at the other point can be calculated by the equation of continuity (Fig. 2.15).

Continuity equation is most commonly used to estimate effective orifice area (EOA) in aortic valve stenosis. Although it is generally reliable, caveats exist.[6–8]

Doppler gradients can overestimate the severity of aortic stenosis in five circumstances, three of which involve errors in measurement.

- Contamination by mitral regurgitation signal
- Dynamic LV obstruction
- Use of angle correction
- High flow states
- Pressure recovery

Limitations of Continuity Equation

While the continuity principle is a fundamental principle of hydrodynamics, the available data for application in a given case have a number of error sources and limitations.

- Pre-stenotic CSA. For the LVOT, it is assumed that its shape is circular and its area can be calculated as πr^2 or $0.785 \times D^2$ (D is the diameter in the parasternal

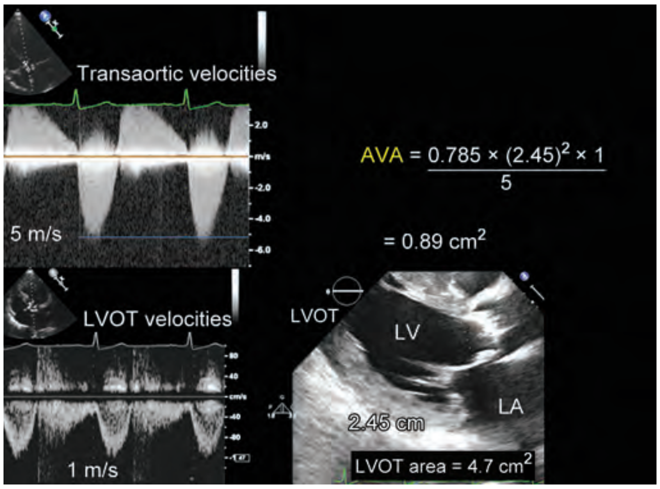

Fig. 2.16: Estimating aortic valve area in a patient with valvular aortic stenosis. The velocity ratio of 1/5 is multiplied by the cross-sectional area of the LVOT assuming it to be circular.

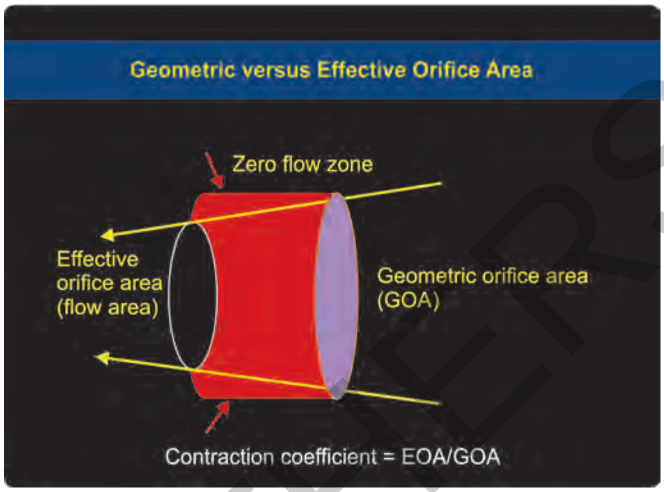

Fig. 2.17: Relationship between effective orifice area (cross-sectional area of flow) and geometric or anatomical area. GOA > EOA by a variable proportion called contraction coefficient.

long-axis view and *r* is the radius or D/2). This is not strictly true, since the outflow tract is in fact elliptical in cross-section, and the parasternal long-axis view shows neither the largest nor the smallest ellipse diameter (Fig. 2.16).

- Furthermore, the true outflow tract diameter is often underestimated if a tangential long-axis cut of the outflow tract is recorded, leading to underestimation of the diameter, radius and ultimately aortic valve area. This is generally regarded as the most important error source in the application of the continuity equation.
- A possible alternative to the full continuity equation, therefore, is the "velocity ratio", which is the ratio of peak velocities or velocity–time integrals in the outflow tract and across the stenosis.
- Pre-stenotic flow velocity and VTI. The flow field in the outflow tract, although laminar, is not homogeneous and blood flows faster at the septal (anterior) side than at the posterior side of the outflow tract (see Fig. 2.2).
- Moreover, velocities by Doppler are often recorded at an angle between flow direction and echo beam, leading to underestimation of velocities.
- Furthermore, in atrial fibrillation, different heart cycle lengths and different preceding heart cycles make it difficult to find equivalent heart cycles for the acquisition of peak velocities or velocity–time integrals in the outflow tract and across the stenotic valve.
- If the maximal velocity across the stenotic orifice is missed by CW Doppler, the continuity equation will produce erroneous results. The continuity equation, therefore, by no means can correct for a suboptimal Doppler interrogation of the aortic valve.

Viscous effects cause lower velocities at the edges of the vena contracta (VC) at low-flow rates, resulting in a parabolic profile. At higher-flow states, inertial forces overcome viscous drag, causing a flatter profile. Effective orifice area itself varies with flow rate as well, with the smallest areas seen at moderate-flow states. These flow-dependent factors lead to flow-rate–dependent errors in the Doppler continuity equation.

The continuity equation calculates an "effective" orifice area, which by definition is smaller than the "geometric" or "anatomic" orifice area (Fig. 2.17).

❒ EFFECTIVE ORIFICE AREA

Effective orifice area (EOA) is determined using echocardiography, Doppler and the continuity equation, and is a reflection of the minimal CSA of the outflow jet (the *vena contracta*). EOA is calculated as the product of the CSA of the LV outflow tract (from its diameter, measured in the parasternal long-axis view) and the LV outflow tract velocity–time integral (using PW Doppler from an apical window), divided by the AV velocity–time integral (using CW Doppler [CWD] from an apical, right parasternal or suprasternal window). Unlike gradients, EOA provides an accurate assessment of stenosis severity independent of flow in most hemodynamic states.

The ratio between geometric orifice area (GOA) and EOA, the "coefficient of contraction", is somewhat

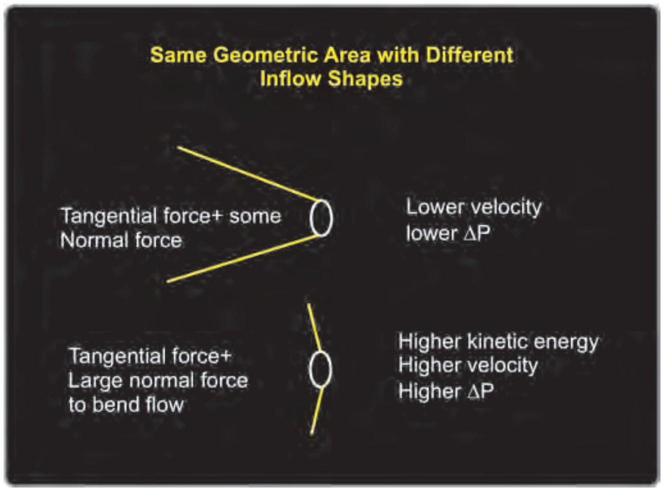

Fig. 2.18: Comparison of funnel-shaped orifice with that of 'hole-in-diaphragm' type with similar geometric orifice areas. Effective orifice area calculated is lower for 'hole-in-diaphragm' type orifice because of higher velocities and ΔP. This is because greater normal acceleration is required to bend the flow into the orifice.

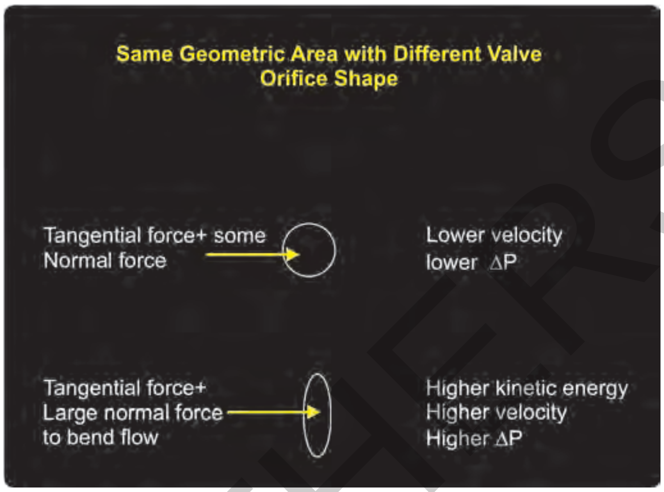

Fig. 2.19: Similar geometric area of two different shapes of the orifice. Elliptical or slit-like shape results in greater ΔP and lower effective orifice area. Contraction coefficient can be as low as 0.60 in such cases.

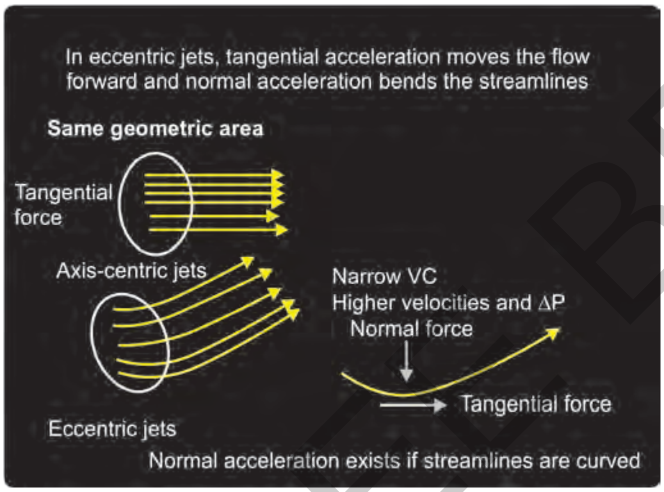

Fig. 2.20: Eccentric jet through an orifice have higher velocities and greater ΔP.

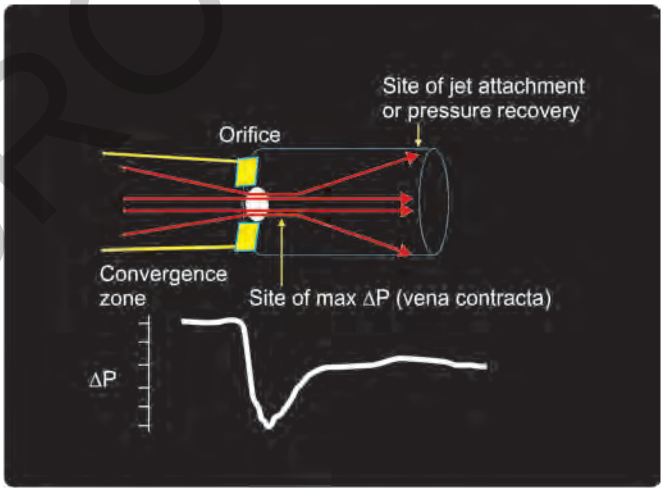

Fig. 2.21: Site of reattachment of the jet to the wall of receiving chamber beyond the orifice.

variable, depending on geometrical features (e.g. smooth or abrupt inlet, Figs 2.18 and 2.19) and also on flow rate, at least if the latter is low.[9] Furthermore, since blood is a viscous fluid, inaccuracies arise from ignoring the boundary layer of the fluid in motion.

Eccentric jets exiting from a narrow orifice also have narrower VC and higher ΔP (Fig. 2.20). This is because of the greater force required to bend the flow.

The EOA and the corresponding pressure gradients are deleteriously affected by increasing jet eccentricity. For the same anatomical size and flow rate, valve morphologies producing eccentric jets have greater energy losses than those that produce concentric jets.[10]

The distance from the VC to the site of pressure recovery also increases with jet eccentricity.[10] Full pressure recovery only occurs when the jet fully expands to fill the aorta, that is, at the most distal reattachment site (Fig. 2.21).

❐ VENA CONTRACTA

It is the point in a fluid stream where the diameter of the stream is the least, and fluid velocity is at its maximum, such as in the case of a stream issuing out of a narrowed valve.[11] The reason for this phenomenon is that fluid streamlines cannot abruptly change direction (Figs 2.22 and 2.23).

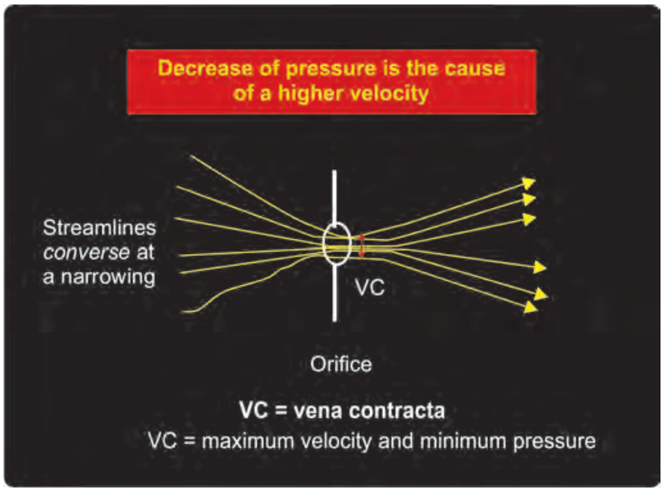

Fig. 2.22: Vena contracta just beyond the narrow orifice, where streamlines converge for a variable distance.

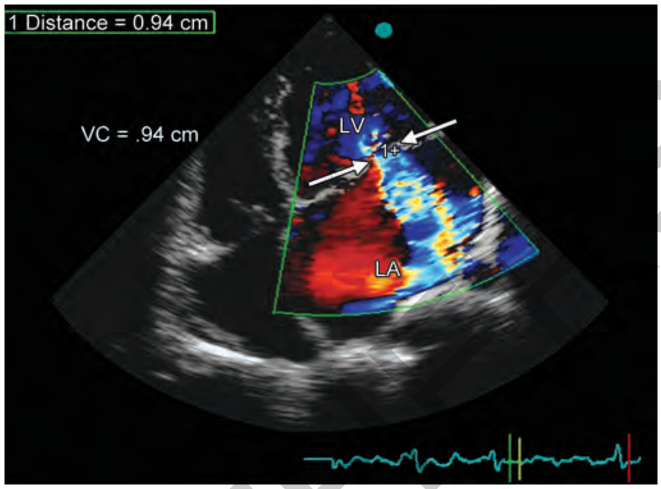

Fig. 2.23: Arrows point to the width of vena contracta of the mitral regurgitation jet.

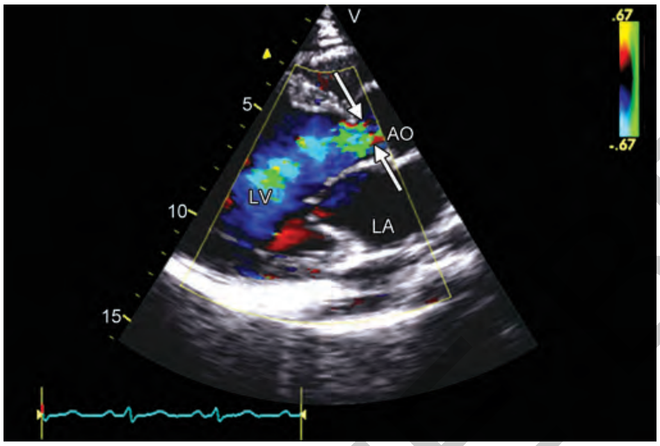

Fig. 2.24: Vena contracta measurement in a patient with aortic regurgitation shown by color flow Doppler in parasternal long-axis view.

The converging streamlines follow a smooth path, which results in the narrowing of the jet (seen by Doppler color flow mapping).

Measurement of the VC is useful as it describes the smallest area of the blood flow jet as it exits a valve. This corresponds to the EOA calculated for valves using the continuity equation (Figs 2.23 and 2.24).

Measurement of diameter of VC is useful in assessing the severity of valvular regurgitations. It is ideally measured in parasternal long-axis views for mitral and aortic regurgitation (AR).[12] Regurgitation is mild if VC ≤ 3 mm and it is severe if VC is ≥ 6 mm.[13]

Even though the measurement of the VC is less dependent on technical factors, small errors in measurement can by multiplied due to the relatively small values of the VC width.

Beyond the VC, the fluid expands, and the velocity and pressure change once again. By measuring the difference in fluid pressure between the normal section and at the VC, the volumetric and mass flow rates can also be obtained from Bernoulli's equation.

It is preferable to use a zoom mode to optimize visualization of the VC and facilitate its measurement. The color flow sector should also be as narrow as possible, with the least depth, to maximize lateral and temporal resolution.

The largest diameter of a clearly defined VC is measured if possible in two orthogonal planes.

Advantages of Vena Contracta

- Simple measure of orifice area
- Valuable in eccentric jets as well
- Not dependent on pulse repetitive frequency
- No correction for angle or convergence walls
- Not affected by other valve involvement
- If the orifice is fixed, then the size of the VC is independent of driving pressure and flow rate.
- Not affected by loading conditions.

Limitations of Vena Contracta

- No temporal information
- May need biplane or triplane measurement to get it right in case of eccentric orifices
- Multiple jets are a problem
- Small errors can make a big difference

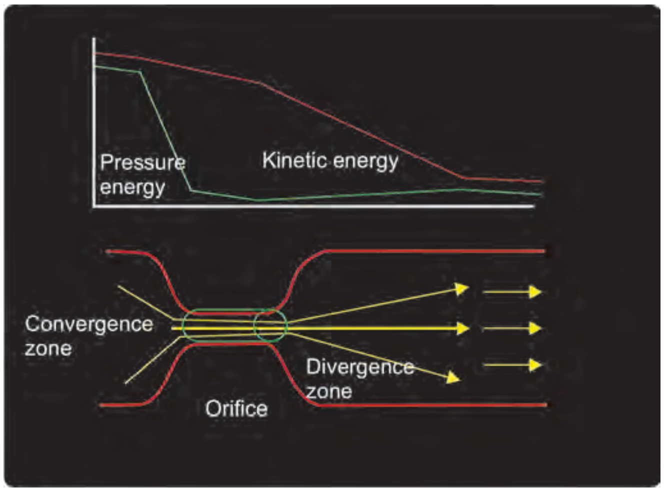

Fig. 2.25: Diagrammatic representation of fluid energy (above) and flow streamlines (below).

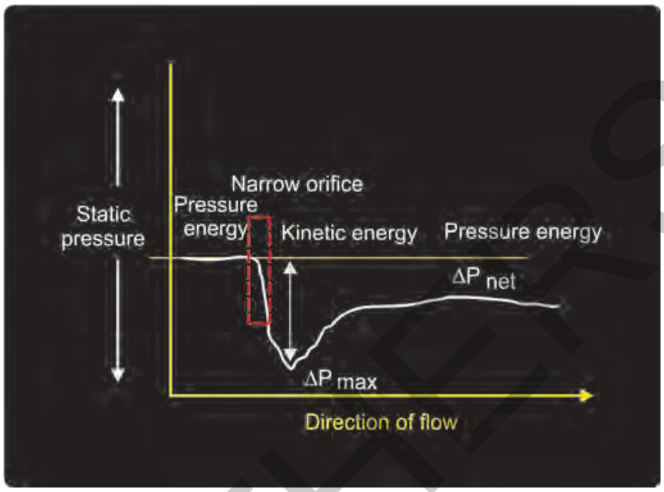

Fig. 2.26: Pressure recovery beyond an orifice is $\Delta P_{max} - \Delta P_{net}$.

- In dynamic regurgitant orifice, vena contracta may change with hemodynamics or during the cardiac cycle.
- The convergence zone is flatter with higher aliasing velocities and become more elliptical with lower aliasing velocities. The aliasing velocity is set between 20 and 40 cm/s.
- Another limitation regards variation in the regurgitant orifice during the cardiac cycle. This is particularly important in mitral valve prolapse, where the regurgitation is often confined to the latter half of systole. The precise location of the regurgitant orifice can be difficult to judge, which may cause an error in the measurement of the proximal isovelocity surface area (PISA) radius.

❐ PRESSURE RECOVERY

Pressure recovery occurs because of re-expansion of flow downstream from the VC of the stenotic jet (Fig. 2.25). When blood moves through a narrow orifice, pressure or potential energy is converted to kinetic energy to drive flow across the orifice. Just beyond the orifice (beyond VC), the streamlines diverge to attach to the wall and reconvert some kinetic energy into pressure energy, resulting in greater pressure drop at the VC, which recovers to some extent as the jet expands fully. This difference in maximum pressure loss to net pressure loss is called pressure recovery (Fig. 2.26).

The pressure recovery phenomenon can be responsible for Doppler gradients that are substantially higher and valve area that is substantially lower than those determined invasively.

Key Points

- At the VC, velocity is a maximum and pressure is a minimum. Beyond VC, pressure recovers to some extent.
- Geometry and turbulence are factors that influence pressure recovery.
- Pressure recovery is reduced by turbulence in the flow field because turbulent dissipation represents irrecoverable energy “loss”.
- There is increased pressure recovery in patients with eccentric orifices, eccentric jets, certain types of bileaflet prostheses and when the receiving chamber is narrow.
- The recovered gradient more accurately reflects the energy “loss” across the valve and, therefore, the left ventricular workload produced by the stenosis.
- Pressure recovery phenomenon can explain high gradients in seemingly normal prosthetic valves (Fig. 2.27) and certain native orificial narrowings (Figs 2.28 and 2.29).

Clinical significance of pressure recovery phenomenon is more in patients with mild or moderate aortic stenosis, who show very high transaortic gradients due to pressure recovery. In these patients, energy loss index can be estimated by measuring aortic diameter (and hence area) at sinotubular junction and using it along with valve area calculated by the continuity equation (Fig. 2.30).

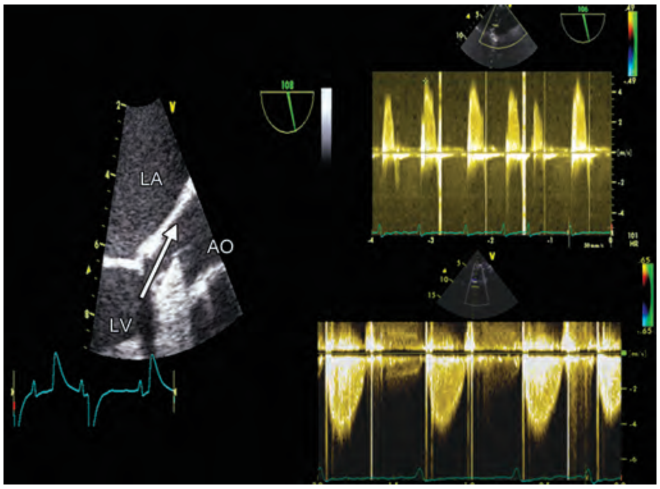

Fig. 2.27: Eccentrically placed Medtronic-Hall valve in aortic position (arrow) showing peak gradient of > 100 mm Hg. Valve motion is normal and the patient is asymptomatic.

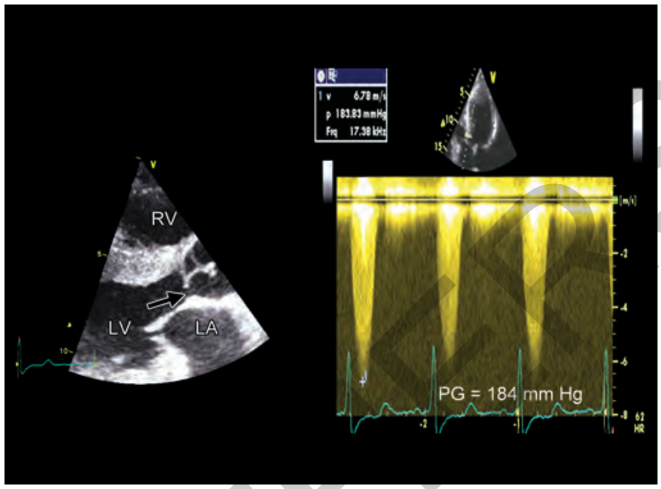

Fig. 2.28: Subaortic stenosis with eccentric orifice in a child with estimated peak gradient of 184 mm Hg. On invasive pressure recording, peak gradient was 80 mm Hg.

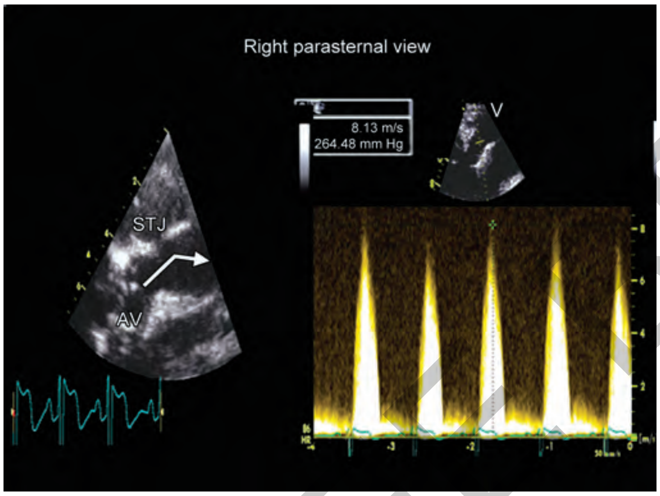

Fig. 2.29: Supravalvular aortic stenosis in a child with eccentric jet. Peak gradient of 264 mm Hg (right panel) is due to highly eccentric orifice and flow.

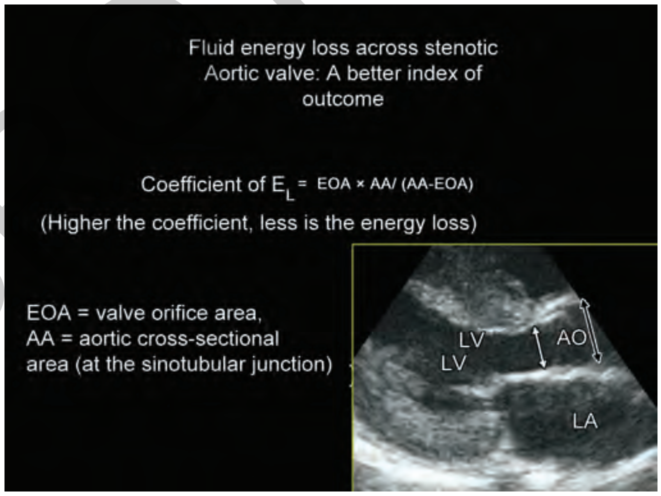

Fig. 2.30: Coefficient of energy loss using a simple formula.

An energy loss index of ≤ 0.5 cm²/M² indicates severe stenosis.[14]

❒ VENTRICULOVALVULAR IMPEDANCE

This Doppler hemodynamic index applies to aortic valve stenosis. It is used to obtain information about the hemodynamic load created by a narrowed orifice. It is a ratio of cuff systolic blood pressure + mean gradient divided by stroke volume index. A ratio > 4.5 to 5.0 suggests higher hemodynamic burden and bad prognosis.[15]

❒ PROXIMAL ISOVELOCITY SURFACE AREA AND VOLUMETRIC MEASUREMENTS

Proximal isovelocity surface area is a Doppler phenomenon to estimate orifice area.[16–18]

- The PISA method is derived from the hydrodynamic principle, which states that as blood approaches a narrow orifice, its velocity increases forming concentric, roughly hemispheric shells of increasing velocity and decreasing surface area (Fig. 2.31).

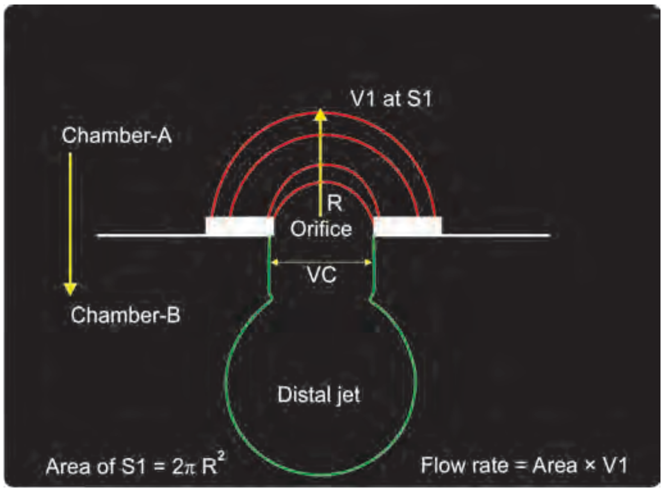

Fig. 2.31: Graphic depiction of flow across a narrow orifice. The red hemispheres represent proximal isovelocity surface area.

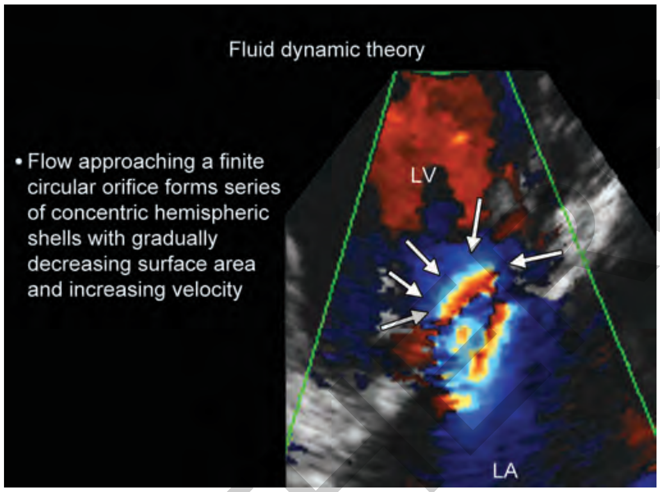

Fig. 2.32: Color flow Doppler of mitral regurgitation showing proximal isovelocity surface area (arrows).

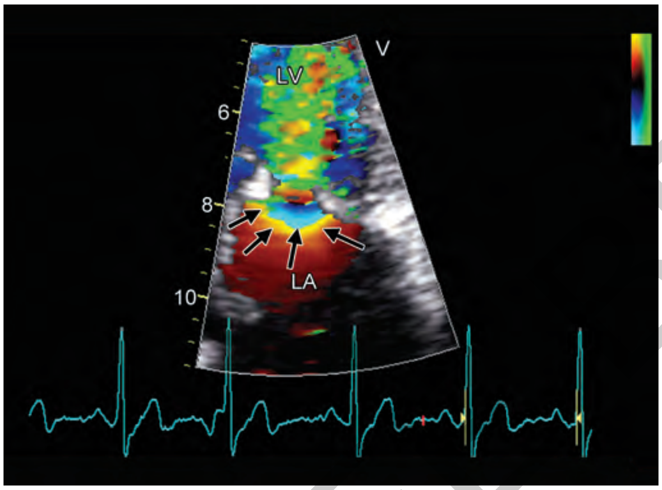

Fig. 2.33: Applying proximal isovelocity surface area method in a patient with mitral stenosis. Note that the orifice does not have a flat surface and small angle correction may be needed.

- Color flow mapping offers the ability to image one of these hemispheres that corresponds to the Nyquist limit of the instrument (Figs 2.32 and 2.33).
- If a Nyquist limit can be chosen at which the flow convergence has a hemispheric shape, flow rate (mL/s) through the orifice is calculated as the product of the surface area of the hemisphere ($2\pi R^2$ or $6.28 \times \text{radius}^2$) and the aliasing velocity (Va) as $2\pi R^2 \times Va$.
- Assuming that the maximal PISA radius occurs at the time of peak flow and peak velocity, the maximal EOA is derived as: $EOA = (6.28R^2 \times Va)/V_{max}$, where V_{max} is the peak velocity of the jet by CWD.
- The transiting volume/beat can be estimated as EOA multiplied by the velocity–time integral of the jet. Since the PISA calculation provides an instantaneous peak flow rate, EOA by this approach is the maximal EOA and may be slightly larger than EOA calculated by other methods. If it is a regurgitant jet, EOA represents regurgitant orifice area.
- Measurement of PISA by color flow mapping requires adjustment of the aliasing velocity such that a well-defined hemisphere is shown. This is generally done by shifting the baseline toward the direction of flow or by lowering the Nyquist limit, or both.
- If the base of the hemisphere is not a flat surface (180°), then correction for wall constraint should be performed, multiplying by the ratio of the angle formed by the walls adjacent to the regurgitant orifice and 180° (Fig. 2.33).

Limitations of PISA Method

- It is more accurate for central jets than for eccentric jets.
- The formula is applicable to a circular orifice. However, in real life, the orifices are more often elliptical or irregular (Figs 2.34 and 2.35).
- If the image resolution allows the flow convergence to be seen well (using zoom mode), and a Nyquist limit can be chosen at which the flow convergence has a hemispheric shape, it is easy to identify the aliasing line of the hemisphere. However, it can be difficult to judge the precise location of the orifice and the flow convergence shape.

A Practical Approach to

CLINICAL ECHOCARDIOGRAPHY

There are already a large number of books on Echocardiography in circulation. Why write another one and increase the confusion amongst the readers? Most books are comprehensive, exhaustive, well-illustrated with chapters written by authorities. These are meant for a select few who are supposed to be skilled operators and crave for enhancing their knowledge. I have tried to read these books but have never been able to complete one chapter in a single sitting. A large number of us are simply busy clinicians like me and want to augment our capability to offer better diagnosis, therapy and prognosis to our patients and for whom learning and practising echocardiography should be fun. For that purpose, there is still scope to bring out a simple book which provides practical tips and can be read at all times. It has been my aim to be practical, simple and to the point with adequate illustrations and images drawn out of my own experience. All single-author books suffer from one drawback. You would see only those clinical conditions which the author has seen. I have avoided mathematical jargon, tedious formulae and equations for which many more books can be consulted. To keep uniformity and principle of simplicity, I have written all chapters myself and labelled all images liberally with illustrations wherever necessary. The book is by no means an exhaustive treatise, and the readers are encouraged to look up the references for detailed information.

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