Equipment in Anaesthesia

An introduction to transoesophageal echocardiography: I. Basic principles

François Béïque MD FRCPC, Denise Joffe MD FRCPC, Simcha Kleiman MD FRCPC

***Purpose**: The purpose of this review is to introduce the uninitiated to transoesophageal echocardiography (TEE): how it works, and what it can do. Sufficient detail is provided to serve as a reference guide to anaesthetists already using TEE in clinical practice.*

***Source**: A Medline search of English language literature up to and including August 1995 was conducted using the key words echocardiography and TEE. Reference echocardiography textbooks were also utilized in the preparation of this review.*

***Principle findings**: All information available from TEE is derived from either cardiac imaging or analysis of blood flow velocity using various Doppler modes. To understand the diagnostic capabilities of TEE we review clinically useful views of the heart as well as modes of cardiac imaging. Sufficient basic physics is presented to allow proper use of adjustment features on the echocardiography machine so that cardiac imaging can be optimized. Available Doppler modes are explained along with an overview of their clinical applications. Figures illustrating clinically useful views obtainable with omniplane TEE are included along with colour prints demonstrating clinical applications of colour flow Doppler.*

***Conclusion**: TEE is becoming increasingly important in the management of cardiac patients for cardiac and non-cardiac surgery. An understanding of the capabilities of the technology as well as the underlying physics allows the anaesthetist to glean the most information from this valuable technique, both quantitatively and qualitatively.*

***Objectif**: L'objectif de cette revue est d'initier le lecteur à l'échocardiographie transoesophagienne (ETO), son fonctionnement et ses possibilités. Les informations incluses sont suffisantes pour servir de texte de référence aux utilisateurs actuels de l'ETO.*

***Source**: Une recherche des publications de langue anglaise s'étendant jusqu'à août 1995 sous les mots clés Echocardiographie et TEE a été effectuée dans Medline. Certains manuels de référence sur l'échocardiographie ont aussi été utilisés dans la préparation de ce texte.*

***Constatations principales**: Toute l'information disponible sur ETO est dérivée de l'imagerie cardiaque ou de l'analyse de la vélocité du flux sanguin avec différents modes Doppler. Pour comprendre les capacités diagnostiques de l'ETO, les vues de base ainsi que les méthodes d'imagerie sont revues. Un rappel de la physique de base, suffisant pour permettre l'exploitation optimale des réglages de l'appareil, est inclus dans le texte. Les divers modes Doppler sont expliqués ainsi qu'un survol de leurs applications. Des figures illustrant les vues de base utiles qu'on peut obtenir avec l'ETO omniplan sont montrées ainsi que des clichés en couleur démontrant les applications du Doppler couleur.*

***Conclusion**: L'ETO joue un rôle essentiel dans l'évaluation et le traitement du patient cardiaque exposé à une chirurgie cardiaque ou autre. Comprendre les limites de cette technologie et la physique sur laquelle elle se base permet à l'anesthésiste d'utiliser au maximum les capacités de cette remarquable technique et de qualifier et quantifier l'information obtenue.*

Key words
ANAESTHESIA: cardiac;
ARTERIES: aorta;
COMPLICATIONS: trauma;
EQUIPMENT: Doppler echocardography;
HEART: echocardiography, myocardial function, heart valves, pericardium, congenital defects;
MEASUREMENT TECHNIQUES: echocardiography.

From the Department of Anaesthesia, Sir Mortimer B. Davis-Jewish General Hospital, McGill University.

Address correspondence to: Dr. François Béïque, Department of Anaesthesia, Room A-335, SMBD-Jewish General Hospital, 3755 Cote Ste. Catherine Road, Montréal, Québec, Canada H3T 1E2

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Transoesophageal echocardiography (TEE) is becoming an important tool used increasingly by anaesthetists in the care of cardiac patients for cardiac and non-cardiac surgery. The advent of this technology has allowed real time anatomical and physiological assessment of cardiac status during the critical perioperative period.

The intention of this review is to make the concepts involved accessible to the uninitiated, while still being complete enough to serve as a reference for those already using TEE in clinical practice. Basic cardiac imaging, as well as different Doppler modes and their uses will be discussed. Sufficient basic physics will be presented to allow an understanding of the technique. We hope this will allow those interested in using the device to gain sufficient accurate information to guide the care of their patients. For others who may not be using TEE in the future, this review will introduce them to the capabilities of the technique.

EQUIPMENT

Echocardiography has long been established as a valuable non invasive diagnostic tool in cardiology and radiology. The development of a transoesophageal M mode (single line imaging) Doppler probe in 1976 met with very limited success since the information was hard to interpret and was of doubtful use for the clinician in the operating room. Ten years later with the advent of the first high resolution 2-dimensional (2D) single plane transoesophageal probe, the potential benefit of TEE became apparent. A single plane transoesophageal probe could be easily inserted into the oesophagus and provide real time 2D imaging of the heart. Today, the probe size is slightly larger than a 9.0 mm endotracheal tube and resembles a gastroscope with a Doppler transducer at the end. Two controls allow for right and left lateral flexion as well as anteflexion and retroflexion of the distal end of the probe to obtain different views of the heart (Figure 1).

The *Single Plane* probe provides a horizontal slice of the heart which is roughly perpendicular to the axis of the oesophagus. Cardiac imaging can be obtained at various levels by withdrawing or advancing the probe in the oesophagus. Anteflexion, retroflexion and lateral flexion of the probe are used to acquire different views of the heart. The probe can be further advanced into the stomach to allow transgastric imaging.

The *Biplane* probe (Figure 2) was the next major advance in transoesophageal imaging technology. This device has two Doppler transducers mounted perpendicular to each other on the tip of the TEE probe. Rotation of the probe is used to acquire most of the images when the vertical plane transducer is used. Biplane technology enables the heart to be viewed in two planes, parallel or perpendicular to the oesophageal axis. It is estimated that the addition of the vertical plane can add as much as 16–44% more information than the single plane.[1,2]

The *Omniplane* probe is the latest TEE probe. A rotating transducer located at the end of the probe can "cut" the heart in all possible planes from 0° to 180° (Figure 2). Since the biplane probe can only obtain views within 20° of each axis, the omniplane probe allows a more comprehensive and easier assessment

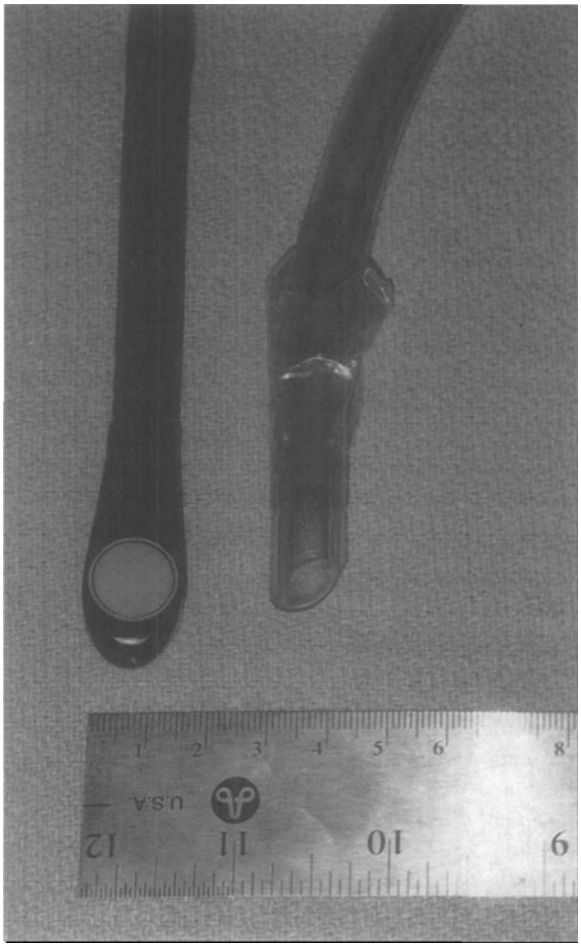

FIGURE 1 The adult omniplane TEE probe is slightly larger than a 9.0 mm endotracheal tube.

than does biplane imaging. Omniplane probes have been shown to have greater diagnostic yield in about 10% of patients and provide more detailed assessment of pathology in 39% of patients.[3]

An *Epivascular* probe is used, in addition to the oesophageal probe, in cardiac surgery. Special high frequency probes are available but, alternatively, a regular transthoracic probe can be employed for epivascular studies. The probe is most commonly used to assess the presence of atherosclerotic disease in the aorta. It is occasionally used to perform an epicardial 2D examination of the heart when intraoperative TEE is not available or is inconclusive. The distal end of the probe is wrapped within a long disposable sterile plastic sheath. After sternotomy, the transducer is positioned directly over the cardiac or vascular structures of interest (*see* Epivascular Imaging).

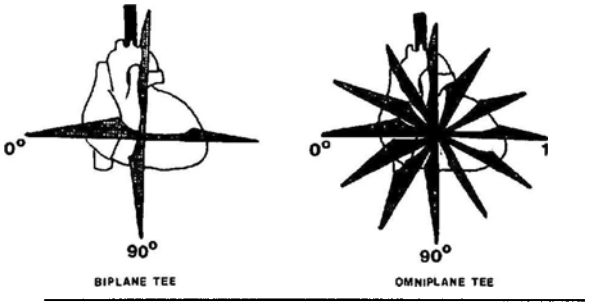

FIGURE 2 Graphical representation of the imaging planes with biplane and omniplane transoesophageal echocardiography.

BASIC PHYSICS

Ultrasound refers to sounds that occur at a frequer beyond the normal hearing range. Ultrasound waves generated by piezoelectric crystals which have unique ability to vibrate when exposed to an electri current and this crystal vibration creates the ultrasou Similarly, when the crystal is exposed to a return ultrasound wave (or a mechanical stress), it will gen ate an electrical current, thus performing the dual role receiving and generating the ultrasound signal. In cli cal echocardiography, the piezoelectric crystals hav very high oscillating frequency, usually from 1 tc MHz. The velocity (v) of the ultrasound wave is eq to the frequency (f) of that wave (determined by the f quency of the transducer) times the wavelength (λ):[4]

$$v = f \times \lambda$$

Resolution of the image is directly proportional to frequency of the piezoelectric crystal. In other wor the more waves that are sent per second, the more def ition can be obtained for any given object. Unfortu ately, higher frequency transducers also have the low tissue penetration. Thus a 2.5 MHz probe is used transthoracic echocardiography (TTE) whereas in TI the heart is in close proximity to the oesophagus a one can use a probe in the 3.7 to 7 MHz ran Therefore, TEE images of the heart usually have a mu higher resolution and appear much clearer than th obtained with TTE.

Most TEE probes allow the operator to choose fr two or three transducer frequencies. One can choose most appropriate transducer setting depending on area of interest. For example, to look at an area close the oesophagus (e.g., vegetations on the mitral valv the best transducer would be a 5 MHz (or 7 MHz available). If one looks at the heart from a transgas view or if it is important to rule out a thrombus at apex from a transoesophageal position, then the lo

frequency setting (e.g., 3.7 MHz) will have more penetration and may allow the best assessment. Another application of this physical principle involves epivascular echocardiography where the transducer is placed directly over the aorta. In this situation a high frequency transducer (e.g., 5 MHz) will provide the best resolution. If there is no epivascular probe available the transthoracic echo probe with the highest frequency (usually 3.5 MHz) should be used.

Attenuation of the ultrasound signal is proportional to the distance travelled but, more important, it will vary depending on the medium through which it passes.[5] Air is the medium with the worst absorption coefficient (Table I) and forms an almost impenetrable barrier to ultrasound signals. Therefore, transthoracic echocardiographic assessment of patients with chronic obstructive pulmonary disease and lung hyperinflation is limited and often inadequate.

TABLE I Half-power distance

Material	*Half-power distance (cm)*
Water	380
Blood	15
Soft tissue (except muscle)	5–1
Muscle	1–0.6
Bone	0.7–0.2
Air	0.08
Lung	0.05

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IMAGING IN ECHOCARDIOGRAPHY

M mode imaging

The first graphic representations in echocardiography were A mode which consisted of a single line of interrogation where each point was represented by lines which varied in length depending on the echogenic density of the medium through which the ultrasound wave travelled, and B mode which showed the A lines end on (Figure 3).[6] These modes are only of historical value and provided the echocardiographer with a still frame of one line across the heart displaying an image which was practically impossible to interpret. The next development was M mode where the dots in B mode are displayed in real time "motion" (Figure 3). In spatial terms, M mode is one dimensional, but because the echo signal is displayed over time, it is really two dimensional. The M mode enjoyed a limited success until it was almost completely replaced by the development of 2D real time imaging. With simultaneous 2D imaging, it became possible to know where that single line was cutting. However, M mode is still used because it provides the highest resolution across a specific line. While 2D image has a frame rate of about 30 Hz, M mode has a frame rate of over a thousand times per second.[6] Therefore, M mode is very useful to analyze events with a temporal component. It is invaluable in the colour Doppler mode to define the timing of an abnormal flow as early, late or pansystolic/diastolic (*see* clinical applications of colour flow Doppler). It also allows precise measurements of changes in cavity size or wall thickness and a close look at wall motion as well as various anatomical structures including vegetations, thrombus and valvular leaflet excursion in real time.

2D imaging

M mode uses a single beam which provides the highest axial resolution. The 2D image is obtained by multiple beams along a single plane. Each beam will generate one line. Older echocardiography machines initially used rotating or mechanical scanners that were followed by linear array transducers to produce the multiple beams (lines) that form the 2D image. These have been replaced with a phased array transducer which consists of multiple transducer elements (usually 48 to 64 in TEE), each with its own circuitry and thin crystal.[7] The firing sequence of these crystals will determine the direction of the final ultrasound wave which will be a summation of all the individual wavelets (Figure 4). Therefore, all the elements or crystals will be used to produce each of the many interrogation beams that will form the final 2D picture we see. As a general rule, the more elements or crystals that are present in the transducer, the better the definition of an individual structure along each line.

The number of lines in a sector, or line density, will determine the overall lateral resolution of the final 2D picture. The frame rate or number of times the picture is updated will determine how smooth the picture will appear. The frame rate depends on the width of the sector and the depth of the sector interrogated. The speed of the ultrasound wave in soft tissue is a constant at approximately 1540 $m \cdot sec^{-1}$.[8] Because the transducer has to wait for the ultrasound wave to return before firing the next wave, the line repetition frequency will depend on the depth (how far the ultrasound beam has to travel) and the width of the sector interrogated (the number of lines which need to be scanned). The speed of the ultrasound wave in soft tissue cannot be changed but the depth and the width of the sector can be limited to the area of interest. Decreasing the depth will allow the machine to spend less time per line. Narrowing the sector to the area of interest will also decrease the time

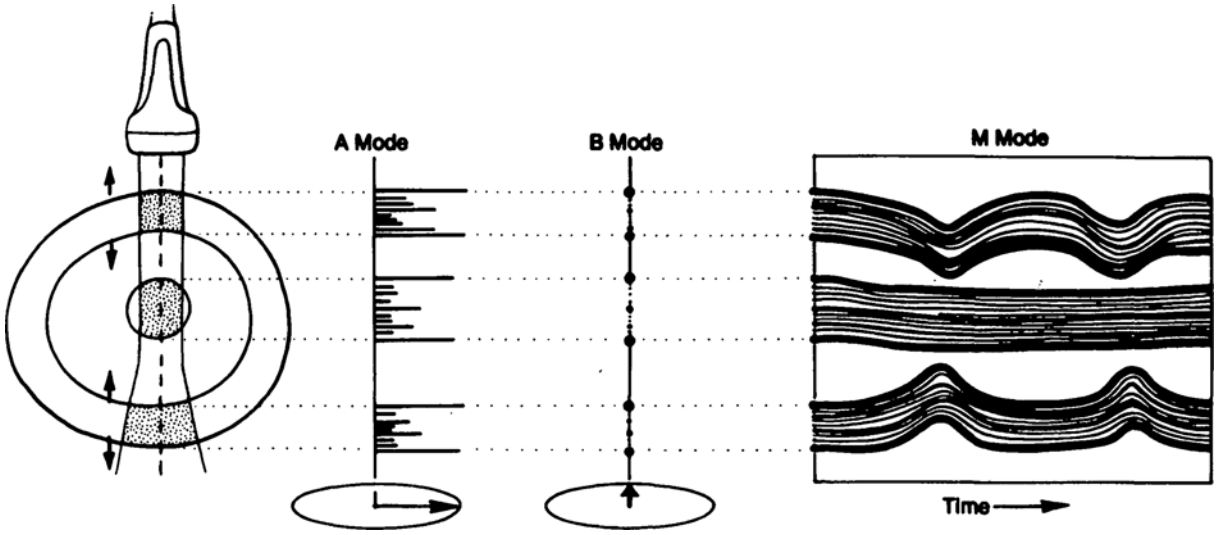

FIGURE 3 Display of echo signals. A mode shows the echoes as horizontal lines, with the amplitude of the signal depicted in the length of the lines. B mode shows the A lines on end represented in dots of varying intensity. M mode shows the sweep of B dots across the television screen or on recording paper. (Reproduced from *Obeid AI*: Echocardiography in clinical practice. Philadelphia, JB Lippincott Company, 1992, with permission).

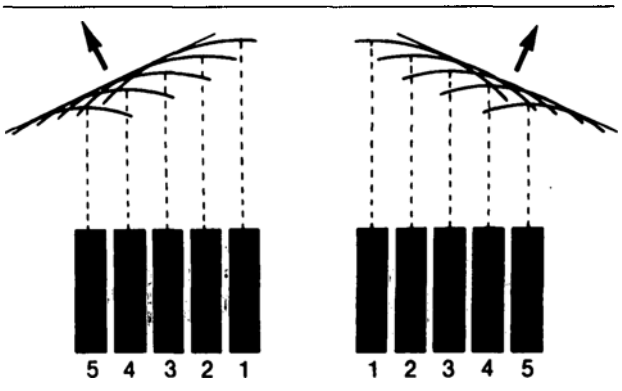

FIGURE 4 Electronic steering of the ultrasound beam in phased array transducers. For simplification, five elements are shown. Beam steering is accomplished by activating the elements slightly out of phase. On the left element # 1 is activated first and element # 5 last. Note how the summation of individual waves from the five crystals results in a wave that is directed to the left. On the right, the order of activation is reversed, resulting in a corresponding change in beam direction. (Reproduced from *Obeid AI*: Echocardiography in clinical practice. Philadelphia, JB Lippincott Company, 1992, with permission).

required to scan that area. The line density (number of lines within a sector) also increases as the size of the sector interrogated decreases. This can considerably improve the resolution of cardiac structures being studied. The net effect of decreasing the size of the sector is a faster frame rate with a better image resolution.

TEE routine views on 2D exam

The order in which structures are examined during a TEE examination is not important as long as it is consistent.[9,10] This helps assure that the examination is complete and that structures aren't missed. Some authors have suggested that viewing commence with the probe in the mid-oesophagus at 0° (Figure 5A). This enables a preliminary assessment of the mitral, tricuspid and aortic valves. The four or five chamber view at this level also provides an estimate of ventricular filling volume and function. Others have suggested initial viewing with the probe in the stomach at 0° (Figure 6A) since only the septal and lateral walls are viewed at 0° with transoesophageal imaging.[5] This allows a more complete initial estimate of ventricular function because it allows visualisation of myocardium that is supplied from all three major coronary arteries (Figure 7).[11,12]

The following is one sequence that can be used during a multiplane TEE examination. An evaluation using a biplane probe can proceed in the same fashion. With the biplane probe, the lateral controls can be used to achieve an additional 20–30° in each plane. A multiplane probe is capable of viewing the heart from 0 to 180°. Therefore, the lateral control knobs are usually not required with omniplane imaging. Only those views that have been found to be clinically useful will be discussed. In addition to cardiac imaging, the echo machine can also measure blood flow velocity using the Doppler modes which are reviewed in the latter section of this article. Views that are commonly used during

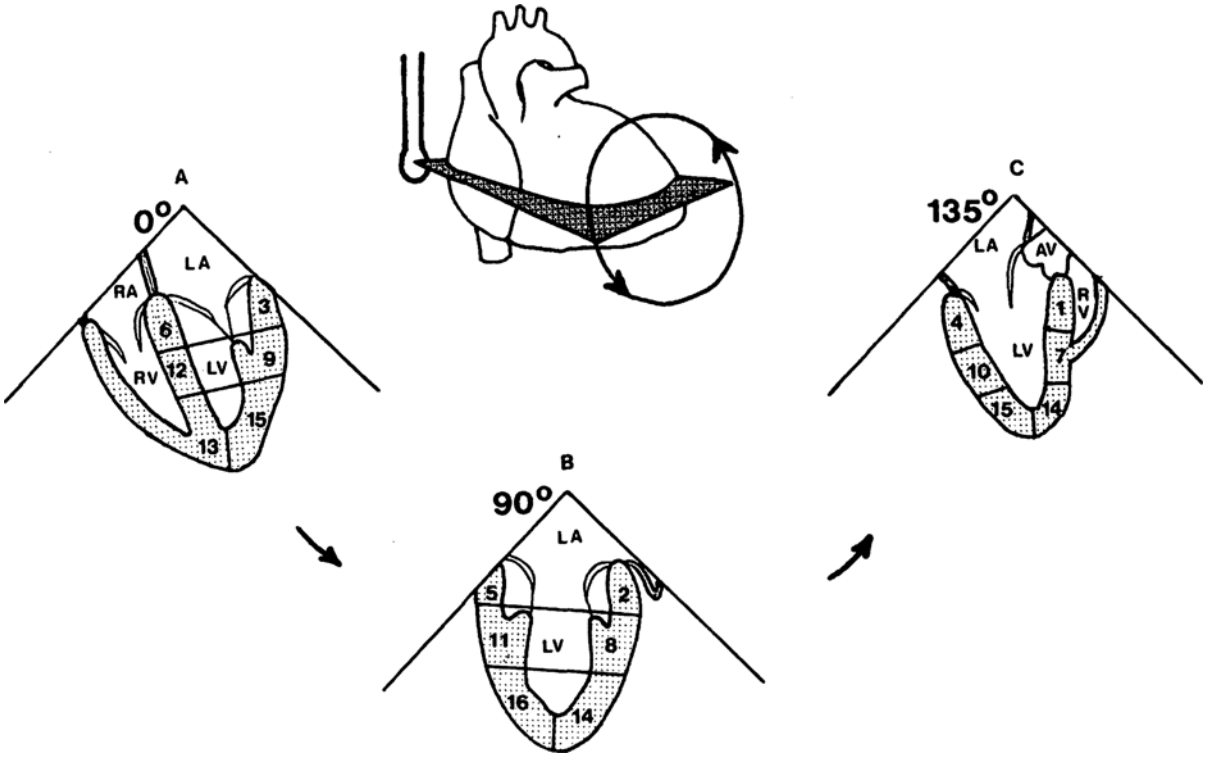

FIGURE 5 Transoesophageal imaging. (A) Mid-oesophageal four chamber view; (B) Mid-oesophageal two chamber view; (C) Mid-oesophageal long axis view. (*Basilar segments*: 1 = Basilar Anteroseptal; 2 = Basilar Anterior; 3 = Basilar Lateral; 4 = Basilar Posterior; 5 = Basilar Inferior; 6 = Basilar Septal. *Mid-segments*: 7 = Mid-Anteroseptal; 8 = Mid-Anterior; 9 = Mid-Lateral; 10 = Mid-Posterior; 11 = Mid-Inferior; 12 = Mid-Septal. *Apical segments*: 13 = Apical Septal; 14 = Apical Anterior; 15 = Apical Lateral; 16 = Apical Inferior. LA = Left Atrium; RA = Right Atrium; LV = Left Ventricle; RV = Right Ventricule; AV = Aortic Valve). (Figures 5A, 5B and 5C were modified from J. Shanewise with permission).

Doppler assessment are therefore also mentioned in the following 2D examination.

Transgastric imaging

Starting with the probe in the stomach at 0° yields the classic short axis mid-papillary view. This view bisects the middle of the heart (the mid left and right ventricles) and demonstrates the anterolateral and posteromedial papillary muscles (Figure 6A). This view has become synonymous with the assessment of left ventricular (LV) function since myocardial regions supplied by all three coronaries are visualized simultaneously (Figure 7). The right ventricle (RV) is seen as a crescentic structure on the left and its function should also be assessed. This view is also used to measure LV dimensions as well. Rotating the transducer to 90° demonstrates the anterior and inferior LV walls from the base to the apex (Figure 6B). The true apex is not always visualized. Rotating further to 120° demonstrates the aortic outflow tract (Figure 6C). Aortic outflow is almost parallel to the Doppler signal in this view which makes it useful for measuring flow velocities.

The transducer is rotated back to 0° and inserted deeper in the stomach. The tip of the probe is anteflexed as much as possible to visualise the heart from the apex (Figure 8). This view demonstrates the aortic outflow. Flow is almost parallel to the Doppler signal and this view provides another opportunity to measure aortic outflow velocities.[13] If the probe is rotated to the right, the right ventricular outflow tract (RVOT) can also be visualized. This view is useful in patients with congenital heart disease to assess the RVOT (e.g., presence of RVOT obstruction).

Transoesophageal imaging

Withdrawing the probe to the lower oesophagus visualizes the mitral valve in short-axis. This image is useful for evaluating the aetiology and location of mitral valvular disease and is especially helpful during mitral valve repair. Further withdrawal of the probe to the mid-oesophageal level brings the right ventricular inflow and coronary sinus into view. If the probe is withdrawn a little more, a four chamber view is seen (Figure 5A). Both atriae and ventricles are visualized along with their

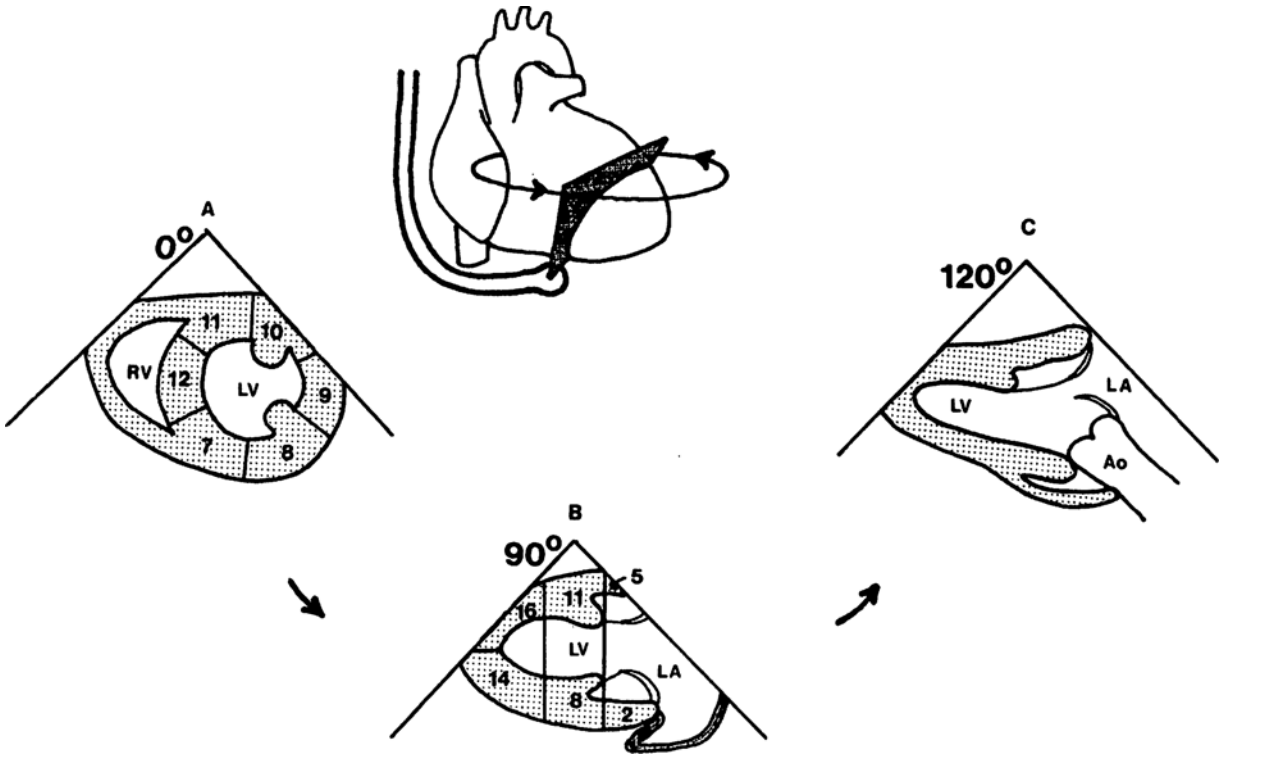

FIGURE 6 Transgastric imaging. (A) Transgastric mid-papillary view; (B) Transgastric two chamber view; (C) Transgastric longitudinal view. Ao = Aorta. (Figures 6A, 6B and 6C were modified from J. Shanewise with permission).

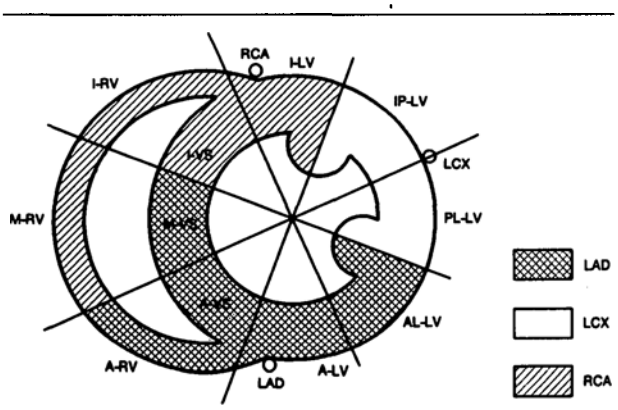

FIGURE 7 Relationship between segments and coronary supplies. LAD = Left Anterior Descending Artery; LCX = Left Circumflex Artery; RCA = Right Coronary Artery; A = Anterior; M = Mid; I = Inferior; IP = Inferoposterior; PL = Posterolateral; AL = Anterolateral; VS = Ventricular Septum; RV = Right Ventricle; LV = Left Ventricle. (Reproduced from *Oka Y, Goldiner PL*: Transesophageal Echocardiography. Philadelphia, JB Lippincott Company, 1992, with permission).

valves. This view is useful for the assessment of ventricular function, valvular function and abnormalities of the atriae. Only the lateral and septal walls of the left ventricle are seen and the true apex is usually not well visualized. Excellent colour and spectral Doppler interrogation of the mitral and tricuspid valves can also be obtained.

Scanning the heart slowly in the four chamber view from 0° to 90°, provides additional information concerning ventricular function. At 90° a two chamber view enables the assessment of the left atrial appendage and of the inferior and anterior LV walls (Figure 5B). This view also allows further assessment of the mitral valvular and subvalvular apparatus. Doppler assessment of left upper pulmonary venous and transmitral flow velocity is ideal in this view. Rotating the transducer to 130–150°, shows the aortic valve and proximal aorta (Figure 5C). This view can usually only be obtained with omniplane TEE and is most useful in the colour Doppler mode to evaluate the severity of aortic insufficiency (AI). However, spectral Doppler assessment of flow velocity is inaccurate in this view because flow is not parallel to the Doppler signal. One must be cautious in evaluating wall motion in this view since the left ventricle may be foreshortened.

Slight withdrawal of the probe from the four chamber view in the mid eosophagus displays the "five chamber" view which includes the aortic outflow tract (the fifth

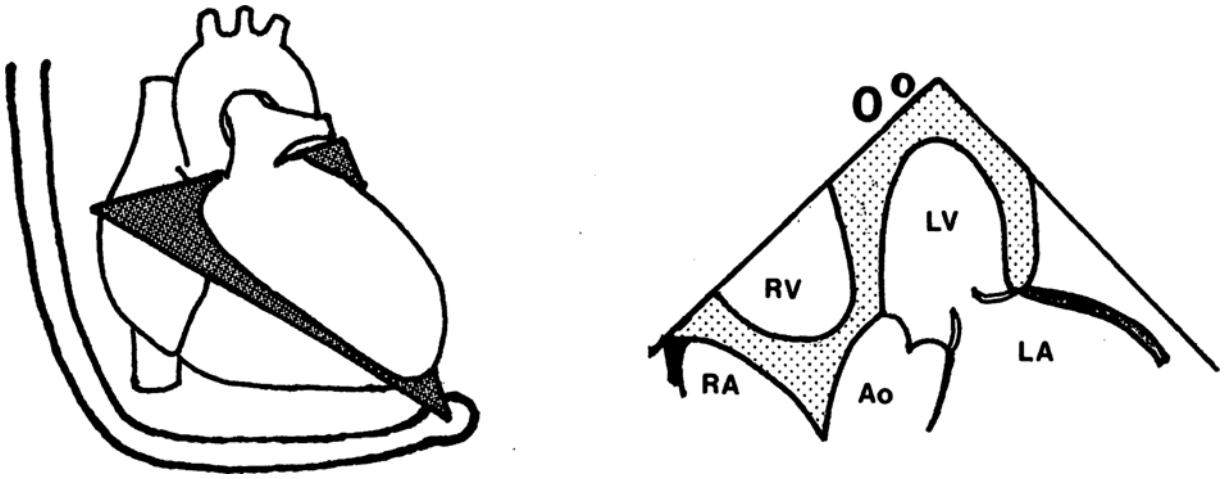

FIGURE 8 Deep transgastric view.

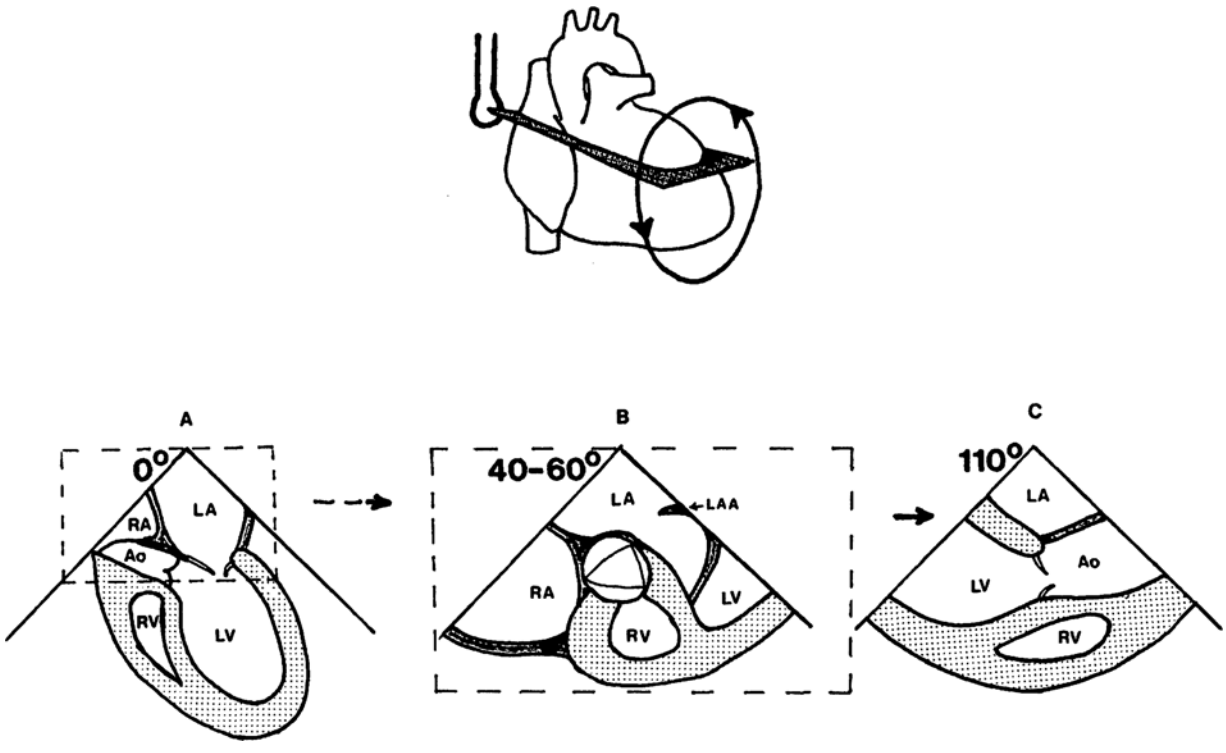

FIGURE 9 Transoesophageal imaging. (A) Mid-oesophageal five chamber view; (B) Mid-oesophageal aortic valve short axis view; (C) Mid-oesophageal long axis view of aorta. LAA = Left Atrial Appendage.

chamber) (Figure 9A). This view is useful for the assessment of aortic valve disease, abnormalities of the LV outflow tract (LVOT) and of the mitral valve. Rotation to 30–60° allows visualization of the "Mercedes Benz" configuration of the aortic valve (Figure 9B). This is excellent for viewing the morphology of the valve (eg. bicuspid versus tricuspid). As the transducer is rotated to 90°, the aorta is unfolded (Figure 9C). These views can also be used to assess the proximal ascending aorta.

The right side of the heart can then be seen by rotating the probe from the mid-oesophageal two chamber

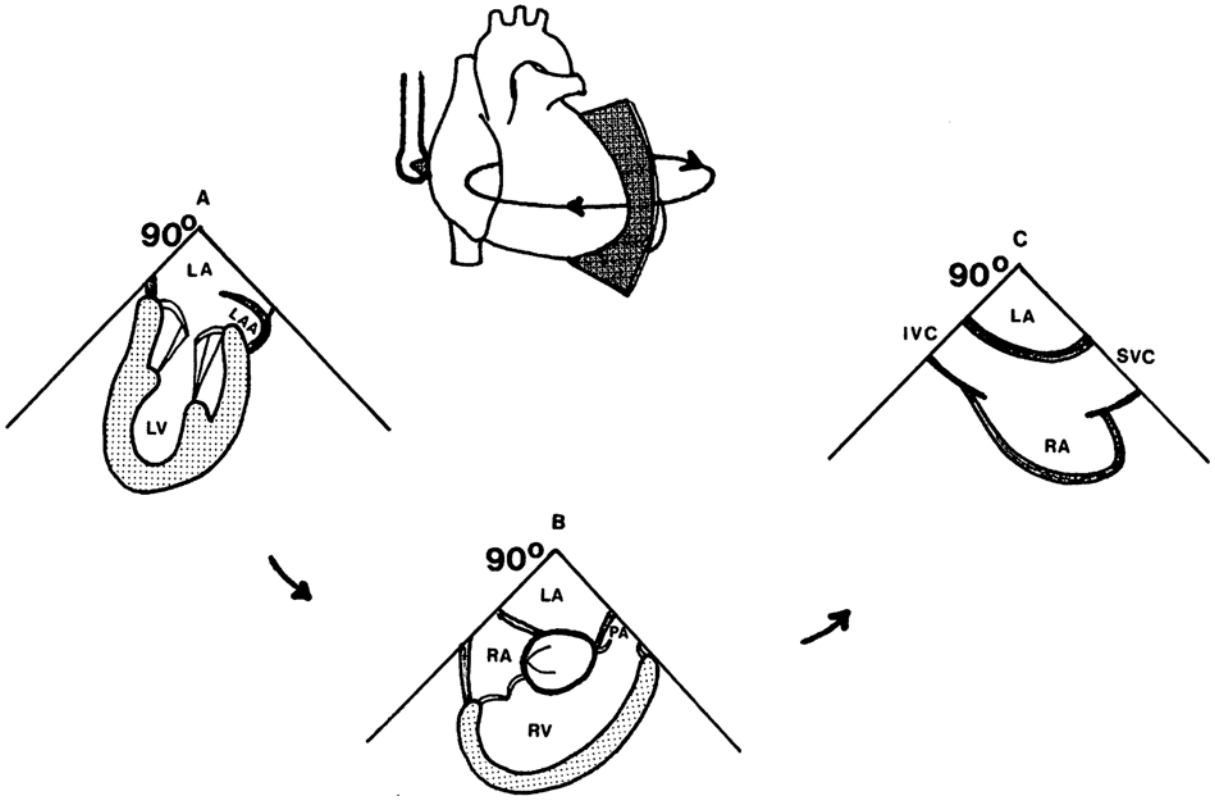

FIGURE 10 Transoesophageal imaging. (A) Mid-oesophageal two chamber view; (B) RV inflow and outflow view; (C) Mid-oesophageal longitudinal RA view. PA = Pulmonary Artery; IVC = Inferior Vena Cava; SVC = Superior Vena Cava.

view towards the right (Figure 10). This will visualize the right ventricular inflow and outflow tracts. Further rightward rotation brings the right atrium (RA), right atrial appendage and vena cavae into view. The interatrial septum and part of the left atrium (LA) can also be seen. This view can be helpful in identifying a patent foramen ovale and in defining types of atrial septal defects (ASD).

The base of the heart is viewed from the upper oesophagus. The main pulmonary artery (PA) bifurcation is seen to the right of the aorta (Figure 11). This view is useful to assess aortic root size, the location of pulmonary artery catheters and abnormalities of the pulmonary arteries.

Finally, the probe is turned towards the patient's back and the thoracic aorta is scanned first in the horizontal plane and then in the vertical plane. At 0° the descending aorta is circular until the arch is reached when it widens. At 90°, the descending aorta is tubular until the arch, at which point it becomes circular.

This examination can generally be completed in 15 min. In cardiac surgical cases, a post-bypass examination need not be as detailed. In general, ventricular function and volume are assessed as well as areas of the heart involved in repairs. It is generally agreed that an exception to the rule of performing a complete examination is when a patient's cardiac instability dictates brevity. In this case a quick look at the structures of interest may be the most practical approach.

Epivascular imaging

Imaging of the ascending aorta for atherosclerotic disease is best achieved intraoperatively with epivascular echocardiography. This technique is predominantly used during cardiac surgery to obtain a detailed assessment of atherosclerotic disease in the ascending aorta and aortic arch prior to aortic cannulation. Embolisation of aortic plaque(s) is probably the most important aetiological factor in the development of perioperative strokes in these patients.[14–17] If aortic atherosclerotic disease can be identified then alterations in surgical technique may decrease the rate of neurological complications.[18,19] Epivascular echocardiography is important because detection of atherosclerotic disease by surgical palpation is accurate in only 50% of cases. Moreover, with TEE, a small portion of the ascending aorta is obscured by the trachea and left main bronchus.[20–22] Unfortunately, this is a major site of surgical manipulation (aortic cannula-

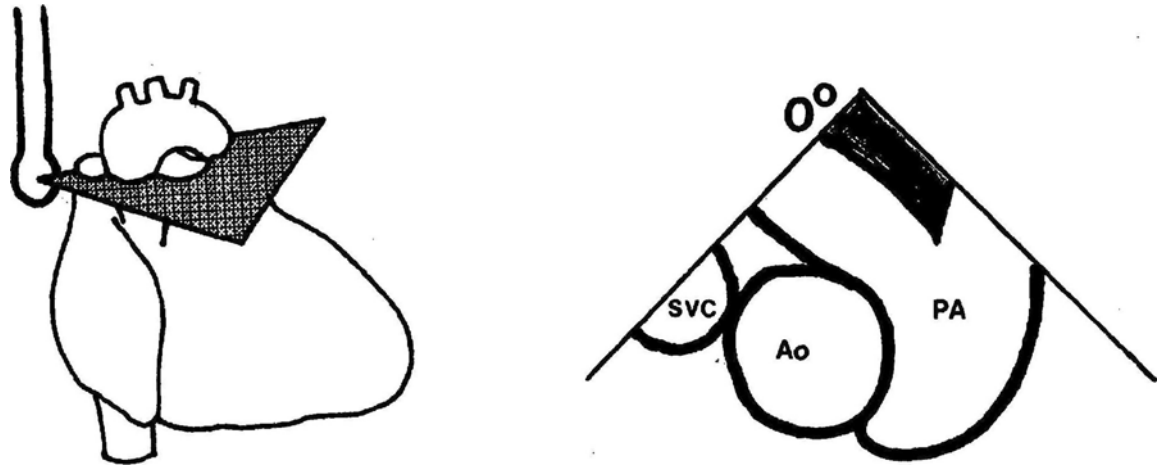

FIGURE 11 Upper oesophageal basal view.

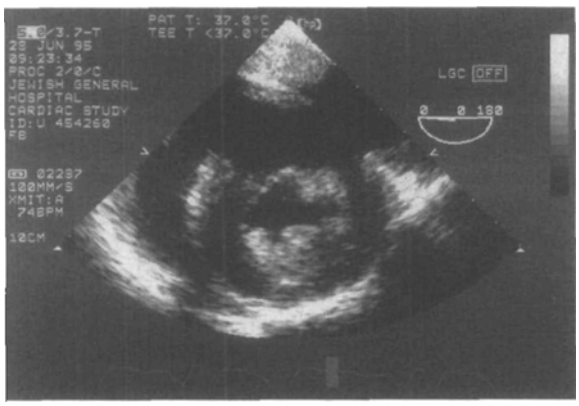

FIGURE 12 Time gain compensation increases or decreases gain in portions of the image.

tion and clamping). Any suspicion of severe disease should be verified with epivascular echocardiography which is the current intraoperative gold standard in imaging of the ascending aorta.[23,24]

The epivascular examination is usually performed by the surgeon. A sterile-wrapped transducer is positioned directly over the aorta before aortic cannulation. The assessment usually starts at the aortic valve and the transducer is slowly moved distally towards the innominate artery. It is important to have a centred circle and not an oblong image of the aorta; this confirms that the transducer is perpendicular to the axis of the aorta. A second assessment can be done with the transducer parallel to the ascending aorta. The posterior wall is well demonstrated with the probe directly over the aorta. To visualize the anterior wall of the aorta where most of the surgical manipulation will occur it is necessary to improve the near field image. This can be achieved by positioning the probe over a water or gel filled glove which is applied to the area being scanned. Alternatively the field can be flooded with warm saline.

2D image processing

The following section reviews several features of the echocardiography machine which are used to optimize the cardiac examination.

*Gain (*power output*)*

The strength of the ultrasound signal will be attenuated as it traverses any medium. The half-power distance refers to the distance where the ultrasound will lose 50% of its original signal through scattering and absorption as it traverses a specific medium. Each medium has its own half-power distance, air being the worst in clinical practice (Table I).[25] Signal attenuation can be compensated for by increasing the overall gain (strength) of the ultrasound signal. Some machines offer lateral gain compensation where the gain can be selectively increased over a specific sector. In addition, most echocardiography machines allow for time gain compensation (Figure 12) where the strength of the ultrasound signal can be increased at a certain depth by modulating the strength of the echo signal returning at a specified time. The moment the echo signal returns will depend on the distance it had to travel which corresponds to the depth of the area of interest. The cost of

increasing the gain or strength of the signal is an increase in noise amplitude and the creation of artifacts.

Dynamic range compression

The ultrasound signal will produce an image of various shades of grey that correspond to ultrasound waves of longer (paler grey) or shorter wavelength (darker grey) depending on the medium traversed. Ultrasound signals within the dynamic range of the echo machine will be displayed on the screen. Very weak signals that are below the sensitivity of the instrument and very strong signals that are beyond the saturation level are automatically eliminated. In clinical echocardiography both strong signals that arise from dense tissues (e.g., cardiac valves, calcifications in the aorta), and weaker signals from soft tissues (e.g., myocardium, endocardium) are important. On a linear scale, many of these signals fall outside the dynamic range and are eliminated. The information is, therefore, processed using a compressed logarithmic scale to increase the resolution and representation at the lower end of the spectrum while accepting stronger signals using a higher saturation level at the other end of the spectrum. It is possible to adjust the level of "logarithmic compression" on most echocardiography machines. Though this sounds complicated, it is really nothing more than adjusting the contrast level on your TV set.

Filter

The filter can be adjusted to eliminate artifacts and "noise." The risk in filtering noise with low and high frequency signals is the elimination of potentially useful information within the filtered range. Previously, only one control was available to increase or decrease the filter level which affected both the low and high frequency signals. Some of the new generation of echo machines have two filters, one for each end of the spectrum, allowing the operator to filter out the low frequency signals or the high frequency signals selectively.

Cine loop

The cine loop is an essential feature on an echo machine. It allows digital acquisition and storage of cardiac cycles which are then displayed on the screen. These images can be reviewed frame by frame later with the ECG tracing to determine the exact timing of any cardiac event.

Depending on the echocardiography machine, the cine loop can be stored as a full screen, split screen, or quad screen. The full screen will have the biggest picture but the split screen (the large screen is divided into two separate screens) and quad screen (four separate screens) allow simultaneous viewing of different recorded images. The split screen is especially useful in the diagnosis of myocardial ischaemia by allowing the clinician to compare a previously recorded image with present real time wall motion.

Zoom

Further definition of cardiac structures can be obtained using the zoom where in addition to limiting the width of the sector, interrogation of the ultrasound beam can also be limited to a specific depth range. When using the zoom feature, a box will appear on the screen which can be moved with the trackball to the area of interest within the 2D picture. The box can be increased or decreased on some echocardiography machines which will then display a very high resolution real time 2D image of that specific area. The smaller the box, the higher the line density and the higher the resolution.

B mode colour

B colour is a function on the echocardiography machine which changes the grey scale of 2D imaging to various hues of blue, purple, orange or yellow. The image is the same but the eye may be better able to differentiate a vegetation, cardiac thrombus, or the endocardium when it is displayed in colours other than grey. B mode colour is also used in spectral Doppler imaging to visualize the velocity envelope better.

UNDERSTANDING DOPPLER MODES

A complete cardiac evaluation includes, in addition to imaging, the use of spectral and colour flow Doppler. The *Doppler shift* is a phenomenon that is encountered every day. For example, there is a characteristic higher pitch sound as a train moves towards us that changes to a lower pitch sound as the train moves away (Figure 13). In echocardiography, a *Doppler shift* occurs when an ultrasound signal is reflected from a red blood cell moving towards or away from the transducer. The *Doppler shift* will be the difference between the frequency of the generated signal and the frequency of the returning signal:[4–6]

Doppler Shift (Fd)

= observed frequency – original frequency

The velocity of the object (in this case a red blood cell) can be calculated from the Doppler shift (Fd) using the following equation:

$$V = \frac{Fd \times \text{speed of sound in soft tissue } (1540\ m \cdot sec^{-1})}{2 \times \text{generated frequency} \times \cos\theta}$$

The generated frequency is the frequency of the Doppler probe which is usually 3.7, 5.0, or 7.0 MHz for the TEE

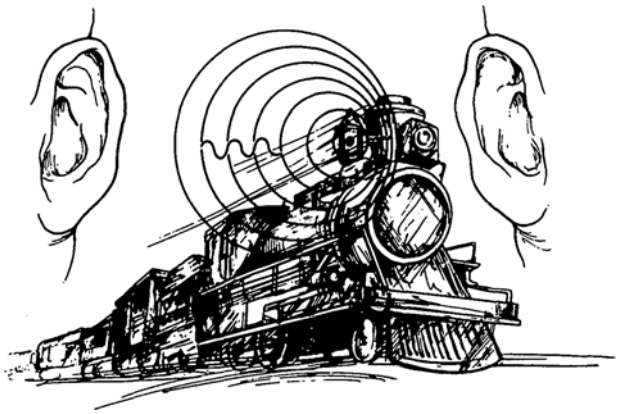

FIGURE 13 The Doppler effect. As the train, which is the source of the emitting sound wave, moves towards the receiving ear, there is a decrease in the wavelength of sound and an increase in the pitch of its whistle. Conversely, when the train moves away from the receiving ear, the wavelength of sound is increased and the frequency of the pitch decreases. (Reproduced from *Giuliani ER, Fuster V, Gersh BJ, et al*: Cardiology: Fundamentals and Practice. 2nd Ed. Mayo Foundation, 1991, with permission).

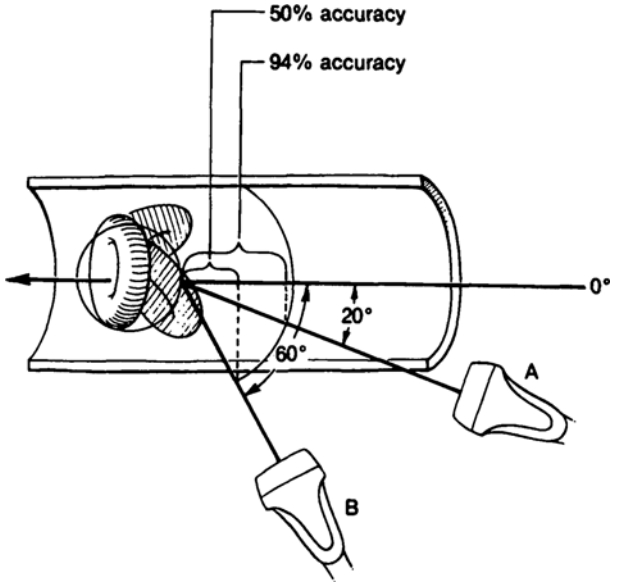

FIGURE 14 Angle of interrogation and accuracy of the amplitude of the Doppler signal. Zero angle occurs with perfect alignment between the flow axis and the interrogating beam. At a 20 degree angle (A) there is only 6% error (94% accuracy), whereas at a 60 degree angle (B) the error is 50%. (Reproduced from *Obeid AI*: Echocardiography in clinical practice. Philadelphia, JB Lippincott Company, 1992, with permission).

probe. To calculate the observed velocity of blood flow, the value is divided by two because the Doppler shift occurs twice, once in the transmitted signal and again in the reflected signal. Blood flow velocity is calculated automatically by the echo machine's software using these parameters.

Theta (θ) represents the angle between the interrogating ultrasound beam and the direction of the measured blood flow. When the Doppler signal is directly in line with the blood flow, θ will have a value of 0, the cos of 0° is 1 and no correction is needed. If θ is perpendicular to blood flow, the cos of 90° is 0 and no velocity will be recorded with either spectral or colour flow Doppler because there will be no Doppler shift. As the angle of interrogation increases, the machine will underestimate flow velocity. Therefore, in spectral Doppler where flow velocity measurements are used to calculate various parameters (*see* Clinical Applications of Spectral Doppler) the interrogating Doppler signal is aligned with the direction of blood flow. An angle >20° should not be used in spectral Doppler for the measurement of flow velocities (Figure 14).[6]

Spectral Doppler

Blood flow velocity can be measured using pulsed wave (PW) or continuous wave (CW) spectral Doppler. Graphic representation of the Doppler shifts obtained with either mode is displayed as a spectral tracing using Fourrier analysis (Figure 15). The amplitude of the spectral Doppler tracing (y-axis) is directly proportional to the measured red blood cell velocity (Doppler shift). The magnitude of the Doppler signal (z-axis) is determined by the number of red blood cells travelling at each of those velocities and is displayed using various shades of grey.[6] By convention, red blood cells moving towards the transducer are represented above the line and red blood cells moving away from the transducer are represented below the line. The patterns of blood flow will show a progression from laminar to turbulent. A well defined envelope with a pale centre is indicative of homogenous laminar flow (Figure 16). Stenotic or regurgitant lesions will be associated with high velocity turbulent flow and spectral broadening because of varied flow velocities and direction.

To understand whether PW or CW Doppler should be used to measure blood flow velocity it is important to know how to acquire the displayed information and its advantages and limitations (Table II).

Spectral pulsed wave Doppler

Pulsed wave (PW) Doppler allows interrogation of flow velocities at specific depths because it will only analyze ultrasound signals returning during a time period which corresponds to the depth of interest (time gating). Although PW Doppler allows for the assessment of flow

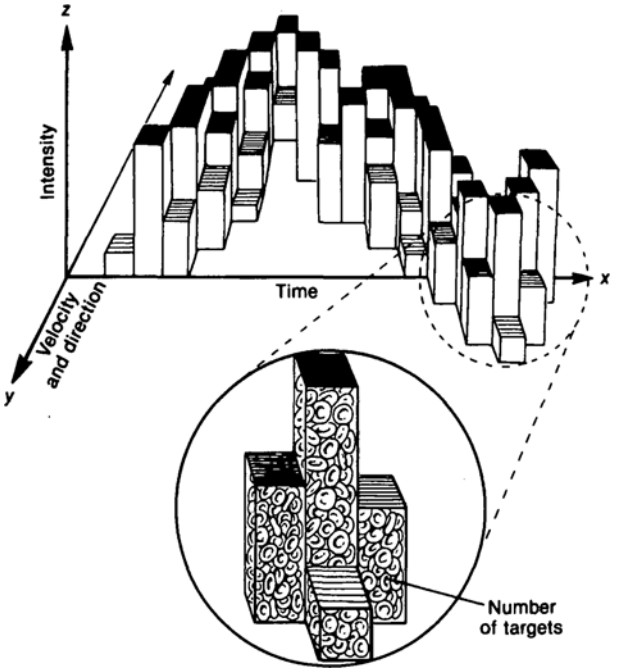

FIGURE 15 Velocity (frequency) vs time, and intensity (amplitude) vs time. Velocity can be displayed as positive or negative, depending on the direction of flow with respect to the transducer. Amplitude, shown by the intensity of the reflected signals, depicts the number of targets. (Modified from *Kisslo J, Adams D, Mark DB*: Basic Doppler Echocardiography. New York, Churchill Livingstone, 1986. Reproduced by permission of Churchill Livingstone, Dr. Joseph Kisslo and Hewlett-Packard Inc.).

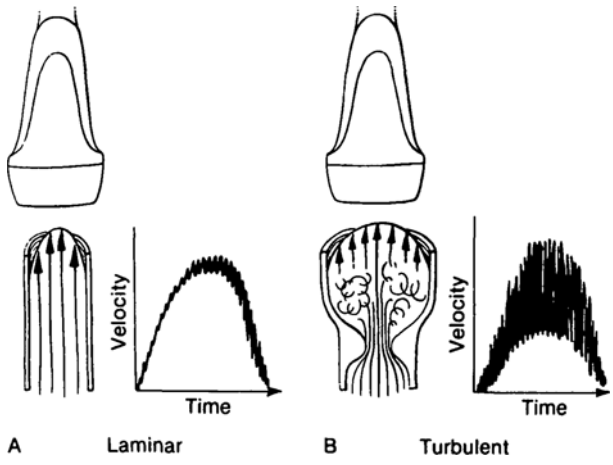

FIGURE 16 Doppler flow profile in laminar (A) and turbulent (B) flow. Notice the spectral broadening of the signal in turbulent flow, representing the inhomogeneity of turbulent flow. (Reproduced from *Obeid AI*: Echocardiography in clinical practice. Philadelphia, JB Lippincott Company, 1992, with permission).

velocity at a very specific location (depth), it cannot measure high velocity turbulent flow (high frequency shifts) because the transducer has to wait for the returning ultrasound signal before sending the next. The transducer alternates between transmission and receiving modes (Figure 17). When PW Doppler is used to interrogate a deeper structure there is a longer delay before the ultrasound wave returns and this will limit the number of pulses generated per second or pulse repetition frequency (PRF). For PW Doppler to measure flow velocity accurately, the PRF of the Doppler signal must be more than twice the frequency of the ultrasound signal generated by the moving object (in this case a red blood cell).[6]

Spectral continuous wave Doppler

Continuous wave Doppler can measure high velocity flows and has an almost infinite PRF because it does not have to wait for the returning signal before sending the next. While in PW Doppler the same transducer listens and transmits, in CW Doppler two different transducers operate simultaneously. One transducer continuously transmits ultrasound signals while the other continuously listens to all returning signals (Figure 17). The disadvantage is that the transducer cannot differentiate which signals come from which depth so that the final analysis will represent an average of all ultrasound signals generated by moving red blood cells along the line of interrogation. In PW Doppler only the ultrasound signals returning from a specific depth along the line of interrogation will be analyzed. This is refered to as time gating.

The Nyquist limit in spectral Doppler

When the PRF is less than half the frequency of the interrogated signal, distortion will occur and the returning signal will "wrap on itself." In echocardiography, this is called *aliasing*. The velocity when aliasing occurs defines the *Nyquist Limit*. The analogy can be found in old western films with a slow image sampling frequency where the wheels of a stage coach appear to be moving backward during obvious forward motion. The Nyquist limit can be increased by increasing the PRF of the transducer. This can be adjusted on most echocardiography machines. However, the Nyquist limit will still be exceeded when measuring high velocity turbulent flows across a stenotic valve with PW Doppler.

In CW Doppler the PRF can be increased almost infinitely since the transducer doesn't have to wait for the returning signal before sending the next. High velocity flows can be measured accurately without aliasing.

Clinical applications

The Doppler shift is used in echocardiography to mea-

TABLE II Pulsed vs continuous wave Doppler

Pulsed wave Doppler	*Continuous wave Doppler*
Same transducer for transmission and for reception	Two transducers: one for each function
Intermittent transmission and reception	Continuous transmission and reception
Limited by pulse repetition frequency: subject to aliasing therefore cannot measure high frequencies	Not limited by pulse repetition frequency: not subject to aliasing and can therefore measure high velocities
Range resolution: present	Range resolution: none
can interrogate blood flow at specific depth with time gating	samples along entire Doppler beam: No time gating

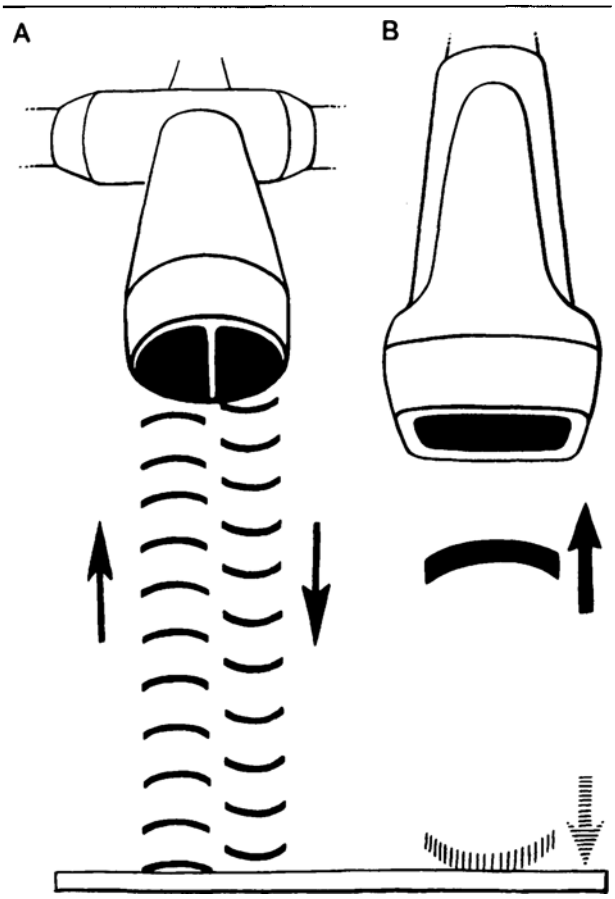

FIGURE 17 Continuous (A) and pulsed (B) Doppler. In continuous wave Doppler, two transducers are mounted together, one for transmission and one for reception. In continuous wave Doppler, two transducers are mounted together, one for transmission and one for reception. In B, only one transducer is employed, which after transmitting a signal waits until it receives the returning echo before it fires again. (Reproduced from *Obeid AI*: Echocardiography in clinical practice. Philadelphia, JB Lippincott Company,1992, with permission).

sure blood flow velocity. This, in turn, can be used to derive several variables useful in quantifying the severity of valvular heart disease. These include pressure gradients and flow velocity ratio across stenotic valves, valvular areas, pressure half-time and regurgitant fraction. Other important applications of spectral doppler include the measurement of cardiac output and the assessment of ventricular diastolic function.[4–6]

MEASUREMENT OF CARDIAC OUTPUT

In the spectral Doppler mode, the measured flow velocities (y-axis) are displayed over time (x-axis). The area under the flow velocity envelope is the velocity time integral (VTI). The product of velocity and time is a distance, therefore the VTI is expressed in units of distance (cm). The VTI is automatically calculated by the echocardiography machine once the flow velocity envelope has been traced. This value is used to calculate stroke volume. The cardiac output can then be calculated once the heart rate is known. On the echocardiography machine, the measurement of the RR interval is used to calculate the heart rate.

$$SV\ (cm^3) = VTI\ (cm) \times \text{orifice area}\ (cm^2)$$

The determination of cardiac output can be done at the level of the mitral valve, the tricuspid valve, the pulmonary artery or the LVOT. The LVOT is used instead of the aortic valve to determine aortic systolic blood flow because it has a circular area which is easier to measure. Flow velocity is measured using PW Doppler and the orifice area can be obtained with 2D echocardiography or spectral Doppler (*see* Valvular Heart Disease).

The area of the LVOT or pulmonary artery can be calculated by measuring the diameter with 2D echocardiography ($\pi \times \text{radius}^2$). Another technique used in 2D echocardiography involves tracing the orifice or valve area to be measured. This is called planimetry and the calculation of the area within the traced border is automatically done by the echocardiography machine.

The presence of aortic insufficiency (AI) or mitral regurgitation will result in overestimation of the cardiac output at those sites since it includes both the forward flow and backward flow across an incompetent valve. Variations in measured cardiac output using this technique may be explained by the fact that orifice area is not necessarily constant or circular throughout all phas-

TABLE III Valvular heart disease

Severity	*Regurgitant fraction*	*Pressure half-time*	*Flow reversal*	*Colour Doppler*	*Mean gradient*	*Valve area*
Insufficiency						
Aortic						
– Mild	< 30%	> 400 msec.		AI < 30% of LVOTD or < 25% of LVOTA		
– Moderate	30–55%	250–400 msec.		AI = 30–60% of LVOTD or 25–60% of LVOTA		
– Severe	> 55%	< 250 msec.	In DT aorta Diastolic	AI > 60% of LVOTD or > 60% of LVOTA		
Mitral						
– Mild	20–30%			MR < 4 cm^2 or < 20% of LA		
– Moderate	30–55%			MR = 4–8 cm^2 or 20–40% of LA		
– Severe	> 50%		In pulmonary veins; Systolic	MR > 8 cm^2 or > 40% of LA		
Stenosis						
Aortic						
– Mild					< 30 mmHg	> 1.5 cm^2 (N = 2.5–3.5 cm^2)
– Moderate					30–50 mmHg	1.0–1.5 cm^2
– Severe					> 50 mmHg	0.8–1.0 cm^2
– Critical					> 50 mmHg	< 0.8 cm^2
Mitral						
– Mild		110–165 msec (N = 20–60 msec)			< 5 mmHg	> 1.5 cm^2 (N = 4 – 6cm^2)
– Moderate		165–220 msec			5–10 mmHg	1.0–1.5 cm^2
– Severe		> 220 msec			> 10 mmHg	< 1.0 cm^2

AI = aortic insufficiency; MR = mitral regurgitation; LVOTA = left ventricular outflow tract area; LVOTD = left ventricular outflow tract diameter; DT = descending thoracic; LA = left atrium; N = normal.

es of systole or diastole. The calculation implies a constant circular orifice area throughout diastole or systole. The transgastric longitudinal view (Figure 6C) and the deep transgastric view (Figure 8) are most commonly used in TEE to determine cardiac output across the LVOT. The four or five chamber view (Figures 5A and 9A) is most commonly used for the determination of CO across the mitral valve (CO calculation across the tricuspid valve has not been validated). Most studies have shown that CO measured at the level of the pulmonary artery using the upper oesophageal basal view (Figure 11) correlates best with thermodilution techniques.[26–29]

VALVULAR HEART DISEASE

Frequent clinical variables which are derived from spectral Doppler measurements of blood flow velocity and which are used to quantify the severity of valvular heart disease are discussed below. The number of techniques used to assess the severity of valvular dysfunction attests to the difficulty that is sometimes encountered. We recommend using more than one technique before grading the severity of valvular lesions (Table III).[30–34]

Regurgitant fraction

The measured forward cardiac output across a regurgitant cardiac valve will be higher because it includes the effective forward cardiac output and the backward regurgitant flow. The difference in measured cardiac output across a competent valve versus an incompetent valve can be used to calculate the regurgitant volume and the regurgitant fraction. The regurgitant fraction may be more useful to quantify the severity of valvular insufficiency because it is more constant over a wide range of cardiac outputs.

In patients with mitral (or tricuspid) regurgitation, the difference between diastolic mitral (or tricuspid) blood flow (VTI × mitral valve area × HR) and aortic systolic blood flow (VTI × aortic valve area × HR) represents the amount of regurgitant blood flow (Table III).[35,36] It is important that the measured CO which is used for

comparison (in this case, the aortic systolic blood flow) does not involve another incompetent cardiac valve. The measurement of cardiac output at any other validated site (*see* cardiac output measurement) can also be used for these calculations:

$$\text{Mitral regurgitant fraction} = \frac{\text{mitral diastolic flow} - \text{aortic systolic flow}}{\text{mitral diastolic flow}}$$

Measurement of the regurgitant fraction can also be used in quantifying the severity of AI[37] (Table III). The difference between mitral diastolic inflow and aortic systolic blood flow represents the regurgitant volume. In this scenario the forward aortic systolic blood flow includes mitral diastolic inflow and aortic diastolic regurgitant flow.

$$\text{Aortic regurgitant fraction} = \frac{\text{aortic systolic flow} - \text{mitral diastolic flow}}{\text{aortic systolic flow}}$$

Calculation of valve area

In echocardiography valvular area can be calculated using the continuity equation. It is based on the premise that total blood flow across competent cardiac valves, the LVOT and the main PA should be equal. The measurement of the aortic valve area will be used to illustrate the practical application of this concept (Table III).[33] Total blood flow across the LVOT should equal total blood flow across the aortic valve (AoV).

$$\text{flow} = \text{velocity} \times \text{area} \quad (\text{area} = \pi \times \text{radius}^2)$$

It has been shown that peak flow velocity (V_{max}) can be used interchangeably with mean blood flow velocity (V_{mean}) in the continuity equation:

$$V_{maxLVOT} \times \text{LVOT area} = V_{maxAoV} \times \text{AoV area}$$

$$\text{AoV area} = \frac{V_{maxLVOT} \times \text{LVOT area}}{V_{maxAoV}}$$

($V_{maxLVOT} = V_{max}$ in the LVOT measured with PW)

($V_{maxAoV} = V_{max}$ across the AoV measured with CW)

The LVOT diameter is measured at the junction of the aortic leaflets with the LVOT. A small error in that measurement may lead to a considerable error in the assessment of the severity of aortic stenosis (AS) because this value is squared to obtain the LVOT area. In this scenario a regurgitant aortic valve would not affect the calculations. If the mitral valve area is to be calculated, total diastolic blood flow across the mitral valve could not be compared to total systolic blood flow across the LVOT if either the mitral or aortic valve were incompetent (*see* Measurement of Cardiac Output). Alternative methods to measure or calculate valvular area with echocardiography are discussed later (see Measurement of Pressure Half-time and 2D Colour Flow Doppler Guided Planimetry).

Measurement of pressure gradients

Measurement of the peak transvalvular pressure gradient across a cardiac valve in echocardiography is done using the Bernoulli principle:[38–42]

$$\text{Peak pressure gradient} = 4\,(V_{max}^2 - V_1^2) = 4V_{max}^2$$

The maximum velocity across a stenotic orifice is measured with CW Doppler. The prestenotic blood flow velocity or V_1, is measured with PW. The measurement of pressure gradients across the aortic valve is shown below to illustrate the application of this principle.

$$\text{Peak pressure gradient} = 4\,V_{max}^2$$

(V_{max} = peak velocity across the aortic valve using CW Doppler).

$$\text{Mean pressure gradient} = 4\,V_{mean}^2$$

(V_{mean} = mean velocity across the aortic valve using CW Doppler).

The square of the maximum flow velocity across a stenotic orifice (V_{max}) is much larger than the square of prestenotic flow (V_1, is usually about one $cm \cdot sec^{-1}$ in the LVOT); therefore this later value is usually excluded from the calculations for simplicity except when the measured V_1, is high (e.g., in the measurement of aortic or pulmonic transvalvular pressure gradients, V_1 will be high in patients with congenital or acquired narrowing of the LVOT or RVOT).

The peak velocity across the aortic valve should be measured with CW for two reasons: first, the peak velocity across a stenotic aortic valve will likely exceed the Nyquist limit and PRF limit inherent to PW mode resulting in *aliasing*; second, only the highest velocity along the line of interrogation is of interest. The mean blood flow velocity or VTI can be obtained by tracing the envelope of the flow velocity signal across the aortic valve in systole.

In aortic stenosis, peak pressure gradients measured by echo differ from that measured by catheterization. In echocardiography, the peak pressure gradient is measured instantaneously whereas, during catheterization, the gradient measured is usually the peak to peak pressure gradient which does not occur at the same time (Figure 18).[41,43] Alignment of the pulsed Doppler signal in the LVOT and across the aortic valve for these measurements can be obtained in the deep transgastric view

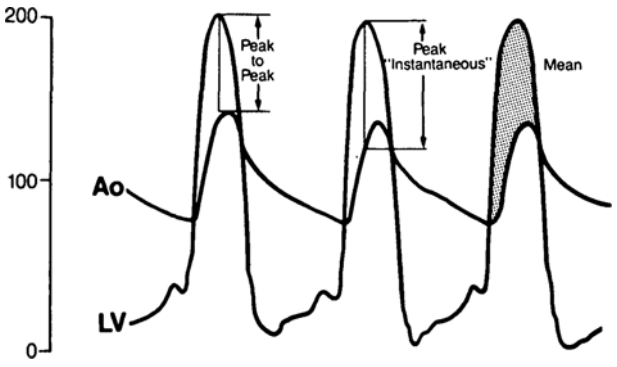

FIGURE 18 The pressure gradient between the aorta (Ao) and left ventricle (LV) has traditionnally been reported at cardiac catheterization as the difference between the peak LV and peak aortic pressures (peak to peak) while Doppler measures the peak instantaneous difference. The mean gradient represents the average pressure gradient throughout systole. (Reproduced from *Labovitz AJ, Williams GA*: Doppler Echocardiography. 3rd Ed. Philadelphia, Lea & Febiger, 1992, with permision).

(Figure 8) and in the 3 chamber long axis transgastric view (Figure 6C). Apparent alignment on the plane of the 2D image represented on the screen may be associated with a large angle between the Doppler signal and the direction of blood flow in other planes.

A more accurate evaluation of the pressure gradient is usually possible during TTE because the Doppler probe is not restricted by the anatomic constraints of the stomach or oesophagus. This is most limiting in the evaluation of aortic stenosis where the measurement of a low pressure gradient with TEE does not rule out the possibility of severe valvular stenosis. On the other hand, the measurement of an elevated pressure gradient is consistent with severe aortic stenosis (Table III). In the evaluation of mitral and tricuspid stenosis the measurement of peak and mean pressure gradient are of limited value because they can vary considerably depending on the heart rate and cardiac output.

Flow velocity ratio

The measurement of pressure gradients are most useful in quantifying the severity of aortic stenosis. Even though these measurements across the aortic valve are less affected by changes in heart rate, a poorly contractile LV may not be able to generate a considerable pressure gradient across a severely stenotic aortic valve. An alternative method which is used more specifically in the assessment of aortic stenosis involves the calculation of the flow velocity ratio which is the ratio of flow velocity in the LVOT (using PW Doppler) to flow velocity across the aortic valve (using CW Doppler). This ratio remains more constant over a wide range of cardiac outputs than pressure gradient measurements. A ratio ≤ 0.25 is associated with severe AS.[43]

Flow velocity ratio

$$= \frac{V_{LVOT} \text{ (PW Doppler measurement in the LVOT)}}{V_{max} \text{ (CW Doppler measurement across the AoV)}}$$

In addition, the flow velocity ratio has been more recently used to assess the severity of mitral stenosis.

Measurement of pressure half-time

Stenotic mitral or tricuspid valvular lesions are associated with a high blood flow velocity during early ventricular filing (*e* wave) followed by a slow decline in velocity and pressure gradient during ventricular diastole. The time required for the pressure gradient to decrease by half during diastole is called the pressure half-time and this reflects the severity of mitral or tricuspid stenosis.[44,45] The pressure half-time depends on the size of the orifice and it remains relatively constant over a wide range of heart rates and flow. The valve area is calculated by the echo machine software from the pressure half-time. In patients with mitral (or tricuspid) stenosis, early passive ventricular filling may be associated with a peak early velocity that may exceed the limit of PW Doppler, therefore, CW Doppler is usually employed for this purpose.

The measurement of pressure half-time is also used in the assessment of valvular insufficiency. The measured blood flow velocity across the aortic valve during diastole represents the instantaneous pressure gradient between the aorta and the LV. The slope and pressure half-time of the AI jet is used to grade the severity of AI (Figure 19).[46] A steep slope suggests a rapidly decreasing aortic pressure (severe AI) or a rapidly increasing LV pressure (severe AI or poor LV compliance). The use of Doppler to assess the severity of AI requires alignment of the CW Doppler signal in the LVOT.

Flow reversal

Severe valvular insufficiency may be associated with a reversal in the normal direction of blood flow at a site distant to the incompetent valve. Flow reversal in the vena cava/hepatic venous system or in the pulmonary venous system is a characteristic of severe tricuspid and mitral insufficiency respectively (Table III). Flow reversal may not be present in all four pulmonary veins especially when the MR jet is eccentric.[47,48] It is therefore important to measure the direction of blood flow in each pulmonary vein to identify systolic venous flow reversal.[43] PW Doppler should be used for this purpose.

In the descending thoracic aorta, the presence of diastolic flow reversal is associated with severe AI (Table

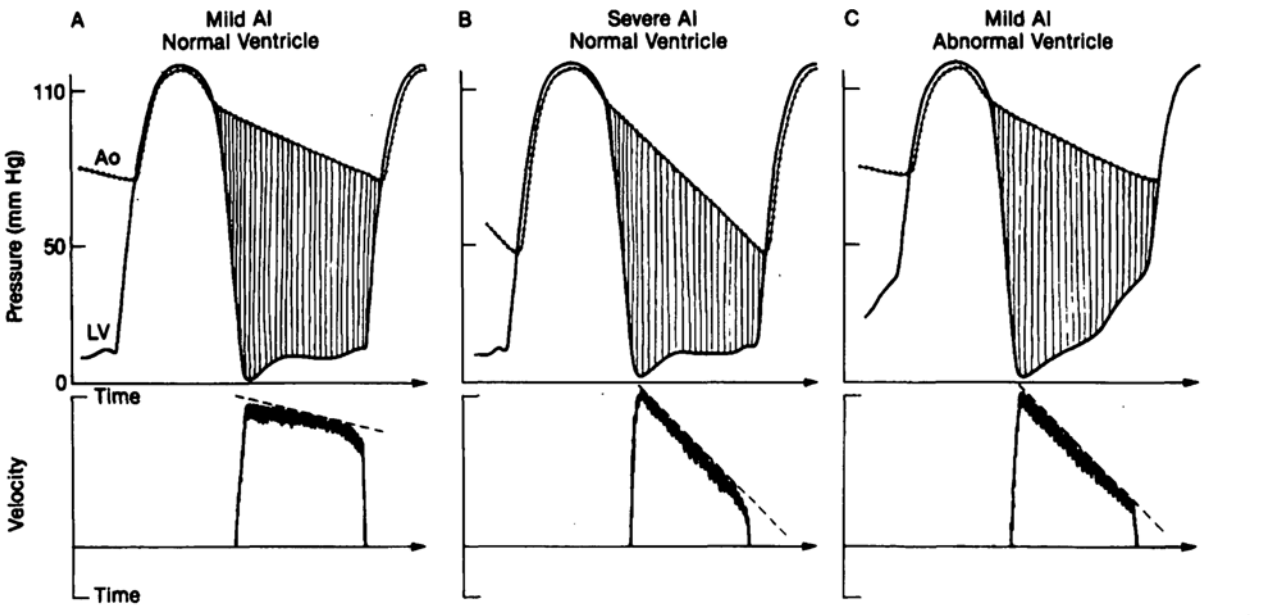

FIGURE 19 Pressure recordings in ascending aorta (Ao) and left ventricle (LV) in upper panel and the corresponding Doppler flow profiles in different settings of aortic insufficiency (AI) in the lower panel. Note that the different slopes of AI envelopes by Doppler are related to the interplay of changes in different clinical settings. (Reproduced from *Obeid AI*: Echocardiography in clinical practice. Philadelphia, JB Lippincott Company, 1992, with permission).

III). It is difficult to align a PW Doppler parallel to blood flow in the descending thoracic aorta because blood flow is almost perpendicular to the Doppler signal.[49] However, It is possible to assess flow reversal in the aortic arch with TEE.

MEASUREMENT OF PULMONARY ARTERY PRESSURES

Pulmonary artery pressures can be estimated in patients who have tricuspid regurgitation. A CW Doppler signal aligned with the jet of tricuspid insufficiency will provide the peak flow velocity (V_{max}) between the RA and the RV. This velocity corresponds to a transvalvular pressure gradient (P_{max}) which can be calculated with the simplified Bernoulli equation:[50,51]

$$P_{max} = 4V^2_{max}$$

The RV systolic pressure is the sum of the transvalvular pressure gradient across the tricuspid valve (P_{max}) and the RA pressure. The peak systolic RV pressure equals the peak pulmonary artery systolic pressure (PAS) provided there is no RVOT obstruction.

PAS = Peak Transvalvular pressure gradient and RA pressure

An arbitrary number can be used for the RA pressure (e.g., 10 mmHg). Alternatively the RA pressure can be measured directly when a central venous pressure line is in place or it can be estimated from jugular venous pulsations. It can also be estimated from the size of the inferior vena cava and its diameter change with respiration.

VENTRICULAR DIASTOLIC FUNCTION

Diastolic function is now considered an important aspect of cardiac function.[52–54] A non compliant ventricle will result in higher central venous pressure or pulmonary artery occlusion pressure for a given volume status. High filling pressures can be misinterpreted as volume overload which may lead to the initiation of inappropriate therapy. Diastolic dysfunction can also result in congestive heart failure even when systolic function is preserved. Ventricular compliance may change abruptly especially in patients with ischaemic heart disease and after coronary artery bypass surgery.[55] Echocardiography allows us to evaluate LV diastolic function with PW Doppler measurements of diastolic blood flow velocity across the mitral valve and in the pulmonary venous system. Normal mitral diastolic blood flow (Figure 20) is characterized by early passive ventricular filling (*e* wave) followed by dynamic ventricular filling with atrial systole (*a* wave). Pulmonary venous flow is characterized by an *S* (systolic) wave and a *D* (diastolic) wave during ventricular systole followed by an *AR* wave (atrial systole) during ventricular dias-

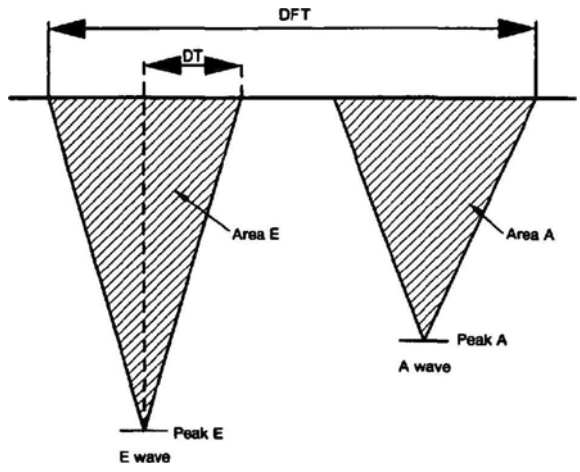

FIGURE 20 Parameters of Doppler measurements. DT = Deceleration Time; DFT = Diastolic Filling Time. (Reproduced from *Oka Y, Goldiner PL:* Transesophageal Echocardiography. Philadelphia, JB Lippincott Company, 1992, with permission).

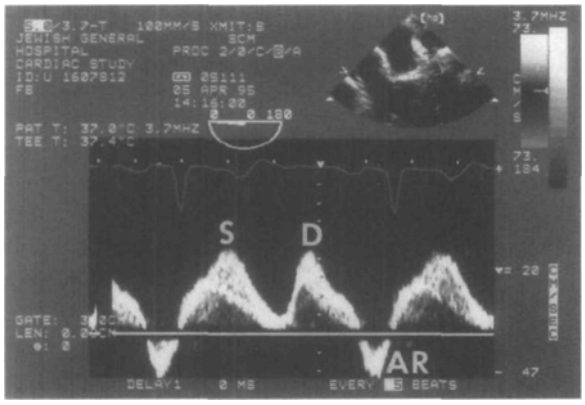

FIGURE 21 Pulsed wave Doppler tracing of left upper pulmonary venous flow.

tole (Figure 21). The *S* wave is caused by downward movement of the mitral valve (descent of the base of the heart) and atrial relaxation which promote passive filling from the pulmonary venous system. The *D* wave occurs with early passive ventricular filling after the mitral valve opens. This is followed by the *AR* wave which represents flow reversal in the pulmonary venous system during atrial systole.[55] Detailed analysis of these waves is beyond the scope of this article. However, cursory analysis allows evaluation of diastolic dysfunction in the absence of mitral valvular pathology. The *e* wave is normally higher than the *a* wave since atrial systole contributes approximately 15–20% of ventricular filling. The *a* wave will usually increase in size with aging which can result in equalization of the *e* wave to *a* wave ratio at age 50.[57]

During Doppler evaluation of mitral diastolic blood flow the *e* wave to *a* wave ratio will vary with small changes in the location of the sampling PW Doppler in relation to the mitral valve. Accurate location of the PW Doppler signal is therefore crucial. It should be positioned at the tip of the opened mitral valve leaflets. Alternatively, the CW Doppler mode can also be used for this purpose and may provide more consistent results when performed by less experienced sonographers. Mild diastolic dysfunction is said to occur when abnormal reversal of the *e* wave to *a* wave ratio is seen. Pseudonormalization of the *e* wave to *a* wave ratio is characteristic of moderate diastolic dysfunction. The clue that pseudonormalization is present can be found in the analysis of pulmonary venous flow. Normally the *S* wave is larger than the *D* wave. A small *S* wave is associated with a non compliant atrial system and high filling pressures.[58,59] The diagnosis of pseudonormalization can therefore be made when the *e* to *a* ratio is normal in the presence of an *S* wave which is smaller than the *D* wave. Severe diastolic dysfunction is associated with a restrictive ventricular filling pattern that consists of an inappropriately high *e* wave to *a* wave ratio with a short isovolumic relaxation time and a rapid early deceleration of the *e* wave (Figure 20). The isovolumic relaxation time is the time period at the end of ventricular systole before the start of the *e* wave. The pattern of severe diastolic dysfunction is usually associated with high filling pressures and a small *S* wave on the pulmonary venous flow Doppler tracing.

COLOUR FLOW DOPPLER

Colour flow Doppler represents a 2D echo image with superimposed colour coding of the Doppler shift created by the moving red blood cells. Colour Doppler uses the same principles found in PW Doppler except that instead of one time gate there are many thousand time gates! In PW Doppler one pulsed signal is sent along a single line while in colour Doppler the transducer sends sound waves in bursts of contiguous pulses. Unlike PW Doppler where the transducer waits for the returning signal before sending the next, in colour Doppler the pulses are sent in succession along each line before the first one has returned. The amount of information that has to be processed is very considerable. Over 250,000 points are analyzed every second and must be colour coded depending on the direction and velocity of the returning signals.[6] Fast Fourrier analysis which is used in spectral Doppler (CW and PW) is insufficient to process all this

information. While PW and CW Doppler can measure peak velocity accurately, colour Doppler only measures mean velocities. The simultaneous use of multiple echocardiographic functions (2D imaging and colour flow Doppler) is called time sharing. Considering the very large amount of data to be processed, it is important to narrow the sector analyzed to the area of interest, otherwise the image on the screen will appear saccadic and in slow motion because the number of frames per seconds will be greatly diminished. If the sector interrogated is too wide and deep, the image may be slowed so much that an abnormal flow that only lasts part of the cardiac cycle could occur and be missed while the machine is still processing the previous information. By convention, blood flow towards the transducer is red and blood flowing away from the transducer is coded blue. In colour Doppler, high velocity blood flow is expressed in brighter colours while darker colour shades are used to represent the lower blood flow velocities.

The Nyquist limit in colour flow Doppler

In colour flow Doppler, high velocity flow will exceed the Nyquist limit, and this is usually displayed as a mosaic colour pattern. Unlike PW where the Nyquist limit is a disadvantage because it limits the peak velocity that can be measured, *aliasing* is clinically very useful in colour flow Doppler. The inability to measure high velocity flow is an advantage in colour flow Doppler because aliasing creates the beautiful and well defined mosaic pattern that is characteristic of turbulent flows seen with shunts, fistulas, defects, valvular insufficiency and stenosis. The area of these abnormal flows can be measured, and in the case of valvular insufficiency, correlates well with the severity of the disease (Table III). As a general rule, it is important to use the highest PRF with colour flow Doppler especially when the heart is hyperdynamic otherwise normal high velocity flow may exceed the Nyquist limit and be displayed with a mosaic pattern resulting in overestimation of regurgitant flow. The exception is in the examination of very low velocity flow as seen, for example, in the false lumen of a dissecting aorta. To determine if there is flow in the lumen, one has to use a very low Nyquist limit (lower PRF).

Clinical applications of colour flow Doppler

2D COLOUR FLOW DOPPLER

Colour flow Doppler is superimposed on the two dimensional image to visualize normal and abnormal flow patterns. In the colour Doppler mode, a mosaic pattern represents abnormal high velocity flows which characterise stenotic and regurgitant valvular lesions (Figure 22). In addition all mechanical prosthetic valves are associated with transvalvular regurgitation. The length, area and direction of these regurgitant jets will vary depending on the type and manufacturer of the valve.[60] A mosaic jet is also usually observed with shunts, fistulas or any abnormal defects or communications.[61]

Two dimensional colour Doppler can also be used to outline poorly visualized lesions on 2D echo. If what appears to be a solid mass (i.e., myxoma or thrombus) is opacified with colour Doppler, then this pseudo-mass is an artifact possibly caused by reverberations. Moreover, an apparent defect with poor image quality can be investigated with colour Doppler to see if there is actually flow across it, or if it is only an artifact caused by an echo drop out. For example, this can occur if the gain setting is too low. Colour flow Doppler is also useful to define the orifice diameter of an abnormal intracardiac communication (e.g., ASD).

2D colour flow Doppler guiding planimetry

Planimetry is done by tracing the opening of an orifice using the trackball. The area within that tracing is then automatically calculated by the software of the echocardiography machine. This technique is most commonly used to assess the severity of valvular insufficiency. Measurement of the maximum mosaic jet area (Table III) and the jet area to atrial area are two techniques that can be used to quantify the severity of mitral insufficiency.[62,63] Even though these measurements are done in the assessment of tricuspid regurgitation, there are no "gold standards" to quantify the severity of the disease. The criteria used to quantify MR cannot be used because the pressures are much lower on the right side. Therefore, a similar mosaic jet area will represent a greater regurgitant volume on the right side. To identify patients who need tricuspid annuloplasty during cardiac surgery, a ratio of maximal right sided jet area to atrial area that is greater than 34% has been used.[64] This measurement has been validated with TTE in the four chamber view. The severity of an eccentric jet will often be underestimated especially if it is "hugging" the wall of the atria (Coanda effect).[65] Colour flow guided planimetry is also useful to quantify the severity of AI. In these patients, the jet area to LVOT area has been shown to correlate with the degree of AI.[66,67] This measurement should be done in short axis immediately below the aortic valve. Planimetry is also used to measure aortic[68] and mitral valve area.[69,70] The addition of colour flow Doppler to 2D imaging can be used to define better the orifice area with planimetry (Figure 23).

PISA with 2D colour flow Doppler

Using colour Doppler imaging of mitral or tricuspid

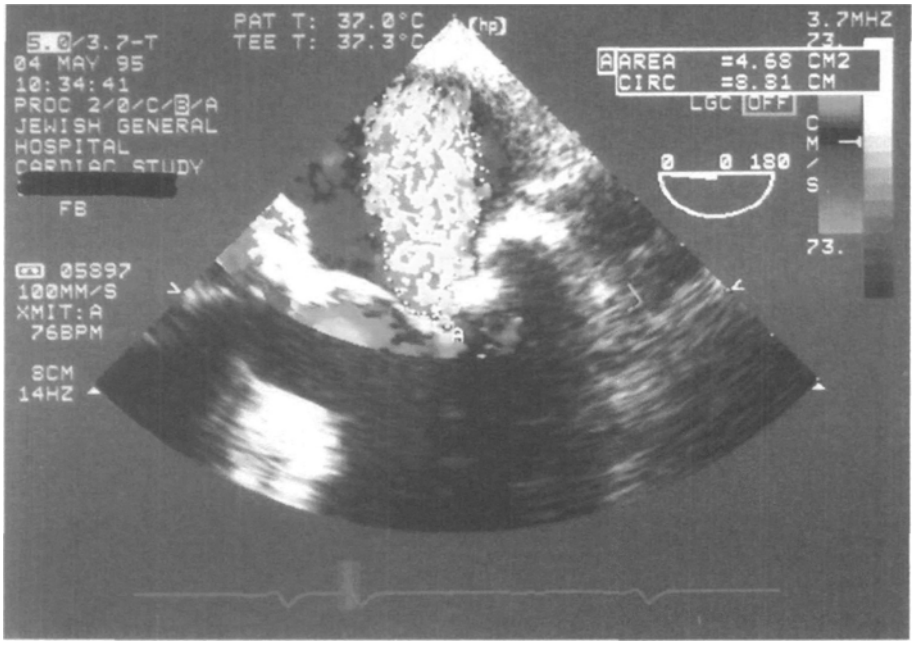

FIGURE 22 A jet of mitral regurgitation (MR) is seen extending into the left atrium in this five chamber view. The MR jet area has been traced.

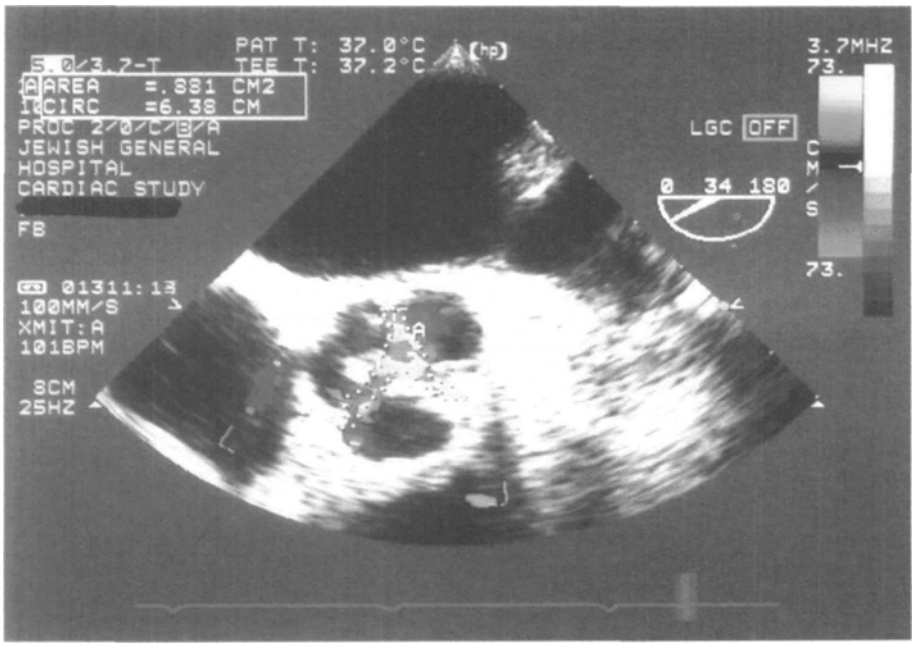

FIGURE 23 Aortic valve area measurement using colour Doppler guided planimetry. (Short axis view of the aortic valve during systole).

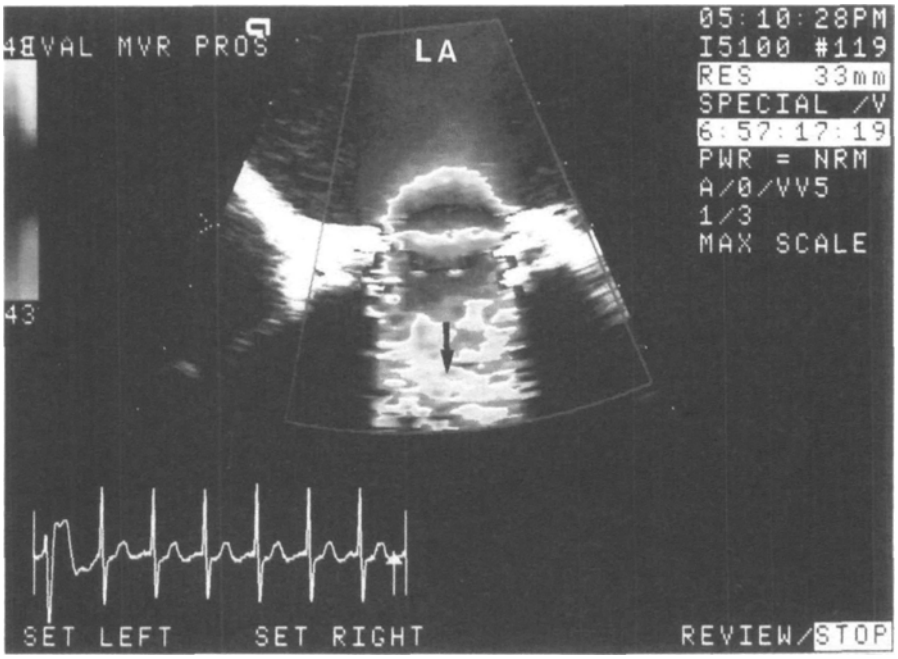

FIGURE 24 Diastolic flow through a porcine mitral valve. Note the proximal isovelocity concentric rings on the left atrial (LA) side of the pro thesis due to relative stenosis and the turbulent flow on the ventricle side of the prosthesis. (Reproduced from *Goldman ME*: Clinical Atlas of Transesophageal Echocardiography. New York, Futura Publishing Company Inc, 1993, with permission).

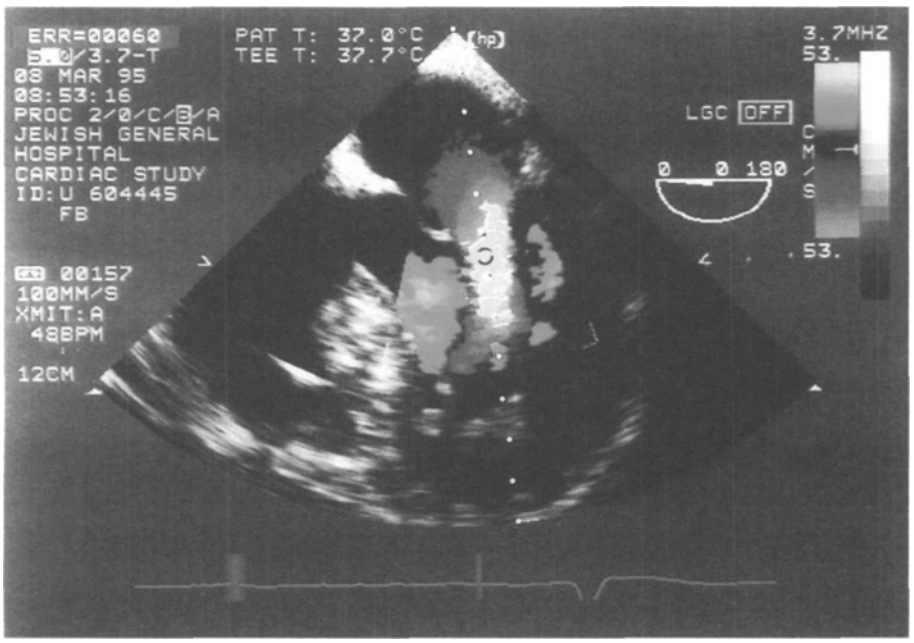

FIGURE 25 In this five chamber view, colour flow Doppler imaging of mitral diastolic inflow is used to align the pulsed wave Doppler signal. Note the change in colour as blood flow velocity increases across the mitral valve. The yellow area represents blood flow velocity which exceeds the Nyquist limit.

stenosis, prestenotic acceleration of blood flow, creates a proximal isovelocity surface area (PISA) which can be identified on the atrial side of the valve during diastole. The height and dome shape characteristic of this diastolic flow may be useful in assessing the severity of valvular stenosis (Figure 24).[71] Measurements of PISA have also been used to quantify the severity of MR and TR.[72,73] In patients with significant MR, the regurgitant flow will be associated with increased systolic flow across the mitral valve causing PISA in the left ventricle. The PISA phenomenon may also be observed in hyperdynamic states across normal valves and can usually be observed in situations where a relatively high blood flow passes through a narrowed orifice (e.g., small ASD).

M MODE COLOUR FLOW DOPPLER

Exact timing and measurement of an abnormal flow can be done with M mode colour Doppler. In hyperdynamic tachycardic patients, it may be difficult to determine if the observed *aliasing* in the LVOT occurs in systole or diastole. In that situation, colour M mode Doppler is very useful to define the exact time period and duration of the turbulent flow in the LVOT. Moreover, in the assessment of AI, the ratio of the width of the regurgitant jet to the width of the LVOT can be measured with M mode colour flow Doppler. This measurement must be done in the LVOT next to the aortic valve and it correlates with the severity of the disease (Table III).[56] It is also very useful in the assessment of aortic dissection to document flow in the false lumen and to confirm the dynamic changes in the size of the false lumen with systole and diastole.

COLOUR FLOW DOPPLER GUIDING SPECTRAL DOPPLER

Another use of colour flow Doppler is to guide CW and PW Doppler measurements. The spectral Doppler cursor is aligned in the area with the strongest colour signal (Figure 25). This enables the most accurate assessment of flow velocity and should be used to guide spectral Doppler measurements.

CONCLUSION

Transoesophageal echocardiography is likely to play an increasing role in the perioperative management of cardiac patients for cardiac and non-cardiac surgery. By understanding not only 2D imaging and Doppler modes, but the physical principles on which they are based, the anaesthetist will be in a position to gain the maximum amount of accurate information from the equipment, while avoiding the creation of a "setting-related" artifact. This will ultimately allow finely tuned care of the sick cardiac patient.

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