

# Principles of Echocardiography

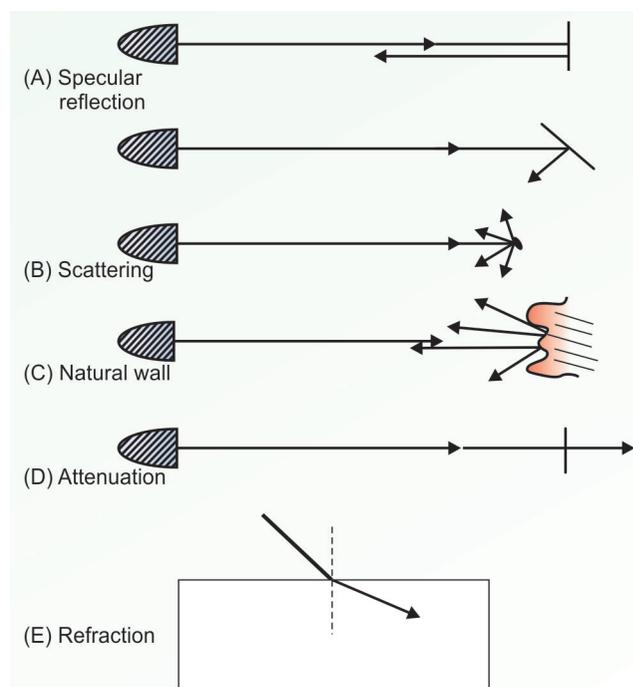
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• Deepak K. Tempe • Suruchi Hasija

**E**chocardiography in the operating room was introduced in the 1970s and the use of transesophageal echocardiography (TEE) during surgery was first described in 1980. Its application grew subsequently with technical developments in high-frequency multi-plane, phased array transducers and color Doppler imaging. Its scope has further widened with the advent of three-dimensional echocardiography and strain echocardiography. The physical principles and instrumentation of TEE involve concepts similar to those of surface echocardiography and are briefly outlined in this chapter.

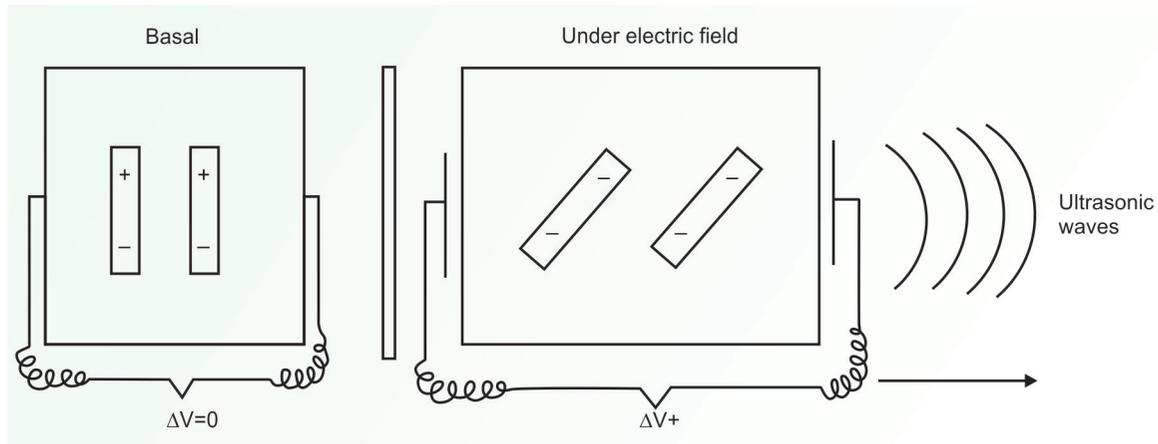
## PHYSICAL PROPERTIES OF ULTRASOUND

Frequency of ultrasonic waves is above the audible range of human ear and exceeds 20 kilohertz (kHz). They can be directed in a beam and obey laws of reflection (Fig. 1.1). At the interface of two media with differing acoustic impedance, ultrasonic waves undergo reflection. Acoustic impedance ( $z$ ) is the density of media ( $r$ ) times the velocity ( $v$ ) of sound in the media ( $z = r \times v$ ). *Specular* or maximum reflection occurs when the angle of incidence is  $90^\circ$ . Incomplete reflection occurs when the angle of incidence is less than  $90^\circ$  (Fig. 1.1A). When an ultrasound beam encounters an irregular surface, the waves are reflected in all directions and that reaching the transducer are proportionately less (Fig. 1.1B, C). Ultrasonic waves are almost completely reflected when the absolute difference in the acoustic impedance of the two interfacing media is large such as soft tissue and metal, calcium, bone or air. The degree of penetration thus varies depending on the acoustic impedance of different media, being relatively poorer for gases and solids. Penetration also depends on the frequency of incidental wave; higher the frequency poorer is the penetration although, the near-resolution is superior. A reduction



**Fig. 1.1:** Patterns of interference between an ultrasound wave and a target. Natural structures are inhomogeneous and ultrasound waves undergo varying degrees of specular reflection (A), scattering (B and C), attenuation (D), and refraction (E).

in resolution occurs with increasing depth and is called attenuation (Fig. 1.1D). Bone and air produce higher attenuation than liquids such as blood and body fluids. Ultrasound waves change direction when they travel from one medium to another. The degree of refraction varies according to the angle of incidence and the difference in the acoustic impedance of the media (Fig. 1.1E). The total thickness that can be traversed by ultrasonic waves is about one-fourth of the wavelength of ultrasound. Since in transesophageal examination, heart is close to the transducer with little intervening



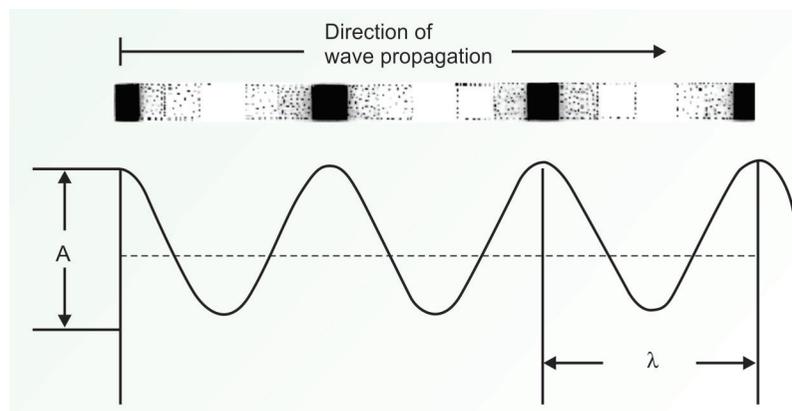
**Fig. 1.2:** Principles of piezoelectric crystals: The crystal generates and receives sound waves using a principle called the piezoelectric (pressure electricity) effect, which was discovered by Pierre and Jacques Curie in 1880. In the probe, there are one or more quartz crystals called piezoelectric crystals. When an electric current is applied to these crystals, they change shape rapidly. The rapid shape changes, or vibrations, of the crystals produce sound waves that travel outward. Conversely, when sound or pressure waves hit the crystals, they emit electrical currents. Therefore, the same crystals can be used to send and receive sound waves.

tissues, the frequency used is relatively higher (3.5–5 million hertz, MHz) than that used in transthoracic examination.

### PRODUCTION OF ULTRASOUND AND TRANSDUCERS

In clinical practice, ultrasonic waves are produced from piezoelectric crystals. These crystals change their shape in electric field and produce alternate contraction and rarefaction of sound waves and conversely they also produce electrical impulses when struck by a sound wave (Fig. 1.2). If a single source (single crystal) were

used, sound waves originating from it would resemble ripples in a pond. When multiple ripples from multiple elements originate, these coalesce to form a unidirectional wave front (Fig. 1.3). Such a wave-front can be advanced in a sector by rotating the elements. This can be done either mechanically or by electric motors. The most popular transducers are electronic real time scanners that use phased array principle. These transducers use multiple elements that are placed linearly and generate a linear wave front. The direction of the wave-front can be altered by delaying sequential activation of the elements. The crystals can be placed



**Fig. 1.3:** Ultrasound propagates through the medium in the shape of a sinuous curve. The permanent transmission of kinetic energy from one molecule to the next is performed in the form of a continuous wave which is referred to as sound wave. During this process, alternating phases of compression (pressure phase) and decompression (suction phase) can be observed in the matter. The maximal height of a wave is the amplitude (A). The intensity of an ultrasound beam is proportional to the square of the amplitude. The ultrasound waves clinically used are low-intensity and do not cause injury. The wavelength ( $\lambda$ ) of a wave is the minimum distance in which a wave repeats itself. The frequency ( $f$ ) of a wave is the number of waves per second. The wavelength and frequency of a sound wave are related as:  $c = f\lambda$ , where  $c$  is the speed of propagation in the medium. As a wave travels from one medium to another, its frequency remains constant. However, the wavelength changes depending upon the speed of propagation.

in sets, one horizontal and other vertical to produce a biplane transducer or a single set can be rotated through 180 degrees as in a multi-plane transducer.

### Resolution

Resolution is the property of an ultrasound beam to discern echoes from sources related in close proximity with respect to space (spatial resolution), time (temporal resolution) or returned signal strength (contrast resolution). The resolution is better with shorter wavelength (depth) and higher frequency of the ultrasound wave.

Spatial resolution is axial, lateral or elevational depending on the ability of the ultrasound machine to identify objects along the axis of the ultrasound beam, horizontal to the beam's orientation or vertical to the beam's orientation, respectively (Fig. 1.4).

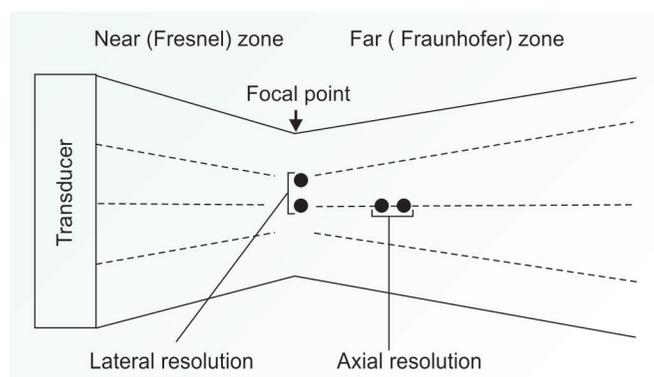
### Variables for Real Time Imaging (Temporal Resolution)

A moving echo image on television screen is made of frames. Usually projection of at least 30 frames/sec is required for eliminating feeling of a stationary image. Each frame is made up of two fields separated by a black line, not appreciated on the screen due to persistence of vision. Each field originates from one ultrasonic sweep. The time required for one sweep is determined by the pulse repetition frequency and sector angle. The rate of emission of ultrasonic waves, which occurs in brief intervals of time (1–2 microseconds), determines the pulse repetition frequency. Similarly the sweep time is also dependent on the sector angle, smaller the sector angle more is the time available for sweeping a given area. The sweep time in turn determines the number of lines in

each sector. An image in ultrasound is made from multiple lines and its resolution depends upon the line density. More the number of lines in one sector, clearer would be the image. Since velocity of ultrasound is 1540 m/sec, each line takes about 0.28 sec to come and fly back. Thus smaller the sector angle, higher would be the number of lines per degree, more would be the line density and higher would be the image resolution. To summarize, the variables that determine an image resolution on screen include line density, pulse repetition frequency, angle of the sweep and the frame rate.

### M-mode and Two-dimensional Imaging

Following emission of ultrasonic waves, the transducer becomes a receiver for the remaining (99%) period and receives the reflected wave. The reflected wave in turn hits the Piezo-electric crystal to produce an electric current. If one knows the time delay then the distance of object can be computed and shown on oscilloscope at a finite representative distance. The amplitude of the returning signal could be represented as a spike (A or amplitude mode) or in the form of varying brightness (B-mode). Subsequent frames over a period of time could be represented on one of the axis to produce motion and called M-mode or motion mode. The use of B-mode image to create exact image of an object in a field-sector and bringing in of subsequent frames with persistence of vision create real time B-mode imaging or the two-dimensional echocardiography. Since the returning signal strength is attenuated due to internal frictional heating, reflection and scattering, ultrasound machines are equipped with time-gain compensation to increase the gain progressively as signals return from deeper tissues. Similarly, the image dropout that occurs when ultrasound beam intersects a surface tangentially is compensated by lateral gain compensation (refer to Chapter 2).

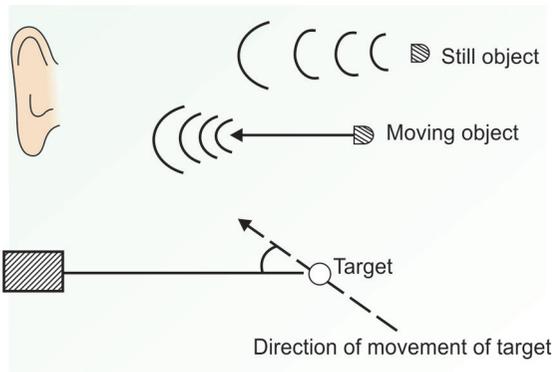


**Fig. 1.4:** The ultrasound wave is transmitted as a beam till the focal point beyond which it diverges like a cone. The lateral resolution is maximal at the focal depth and decreases in the near (Fresnel) zone and the far (Fraunhofer) zone.

### Doppler Echocardiography

Doppler is a technique to detect the manner in which blood moves in the cardiovascular system. If the target is stationary, the frequency of transmitted wave and reflected wave is identical. If the target is moving towards the transducer, the received frequency is increased. If it moves away, then the frequency is decreased. The Doppler shift represents the difference between received and transmitted frequencies (Fig. 1.5). The mathematical relation between the velocity ( $v$ ) of the target and Doppler frequency can be given by the following equation

$$\text{Doppler equation: } v = \frac{fd \cdot c}{2 \cdot ft \cdot (\cos \theta)}$$

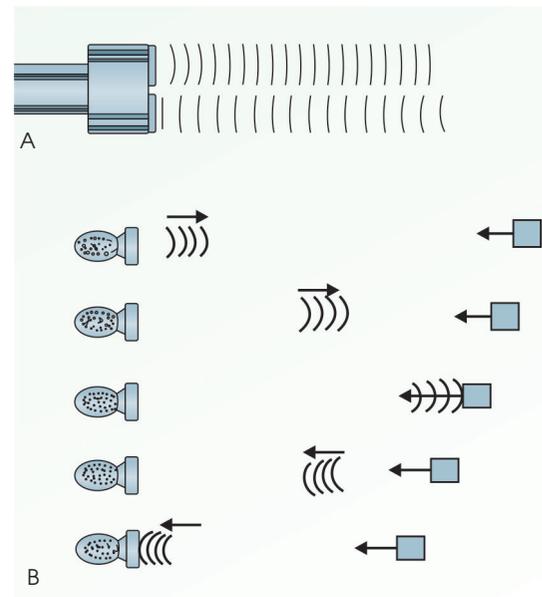


**Fig. 1.5:** Doppler effect. Sound waves that are emitted from an object moving towards a receiver are compressed causing an increase in the frequency.

where  $fd$  is the Doppler frequency,  $ft$  is the transmitted frequency,  $c$  is the speed of propagation in the medium and  $\theta$  is the angle between path of travel and ultrasonic beam.

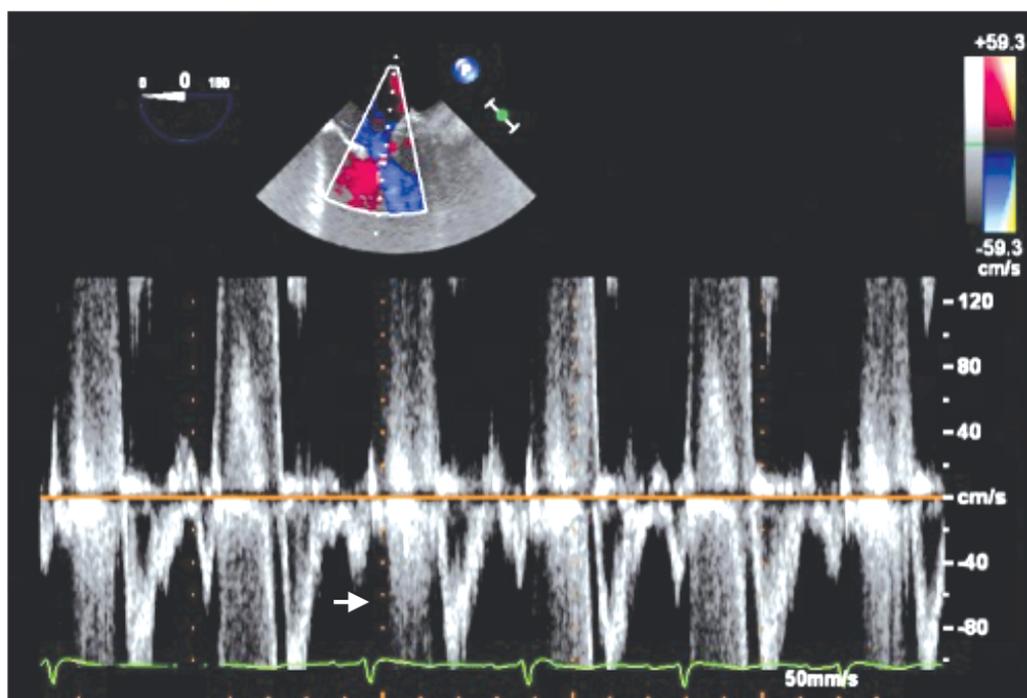
The best Doppler information is derived if the beam is parallel to moving target with an angle between the beam and direction of moving target not exceeding 20 degrees. This is paradoxical to 2-D echocardiography wherein the best information is derived if the beam is perpendicular to the region of interest. Two types of Doppler application are used in clinical practice: continuous wave and pulsed wave Doppler (Fig. 1.6). In continuous wave Doppler, a transducer is used with separate transmitting and receiving elements. This works continuously and hence there is no way to register the delay in receiving an incoming impulse, and thus the depth of an individual target or the location of a moving target cannot be estimated. On the contrary, in the pulsed wave Doppler, the same transducer initially sends a burst of waves and then works as a receiver. By knowing the delay, it is possible to compute the distance of a moving target or to obtain information from a given depth. However, the major limitation of pulsed wave Doppler is that it cannot measure velocities above a specific threshold. Since a finite time is required for transmitting a returning wave, frequency higher than a specific threshold cannot be registered. This is called an aliasing velocity and the limit is called Nyquist limit (Fig. 1.7). Usually the Nyquist limit equals one-half pulse repetition frequency.

The Doppler information is usually displayed graphically against time and is called the spectral Doppler. Use of spectral Doppler is useful for

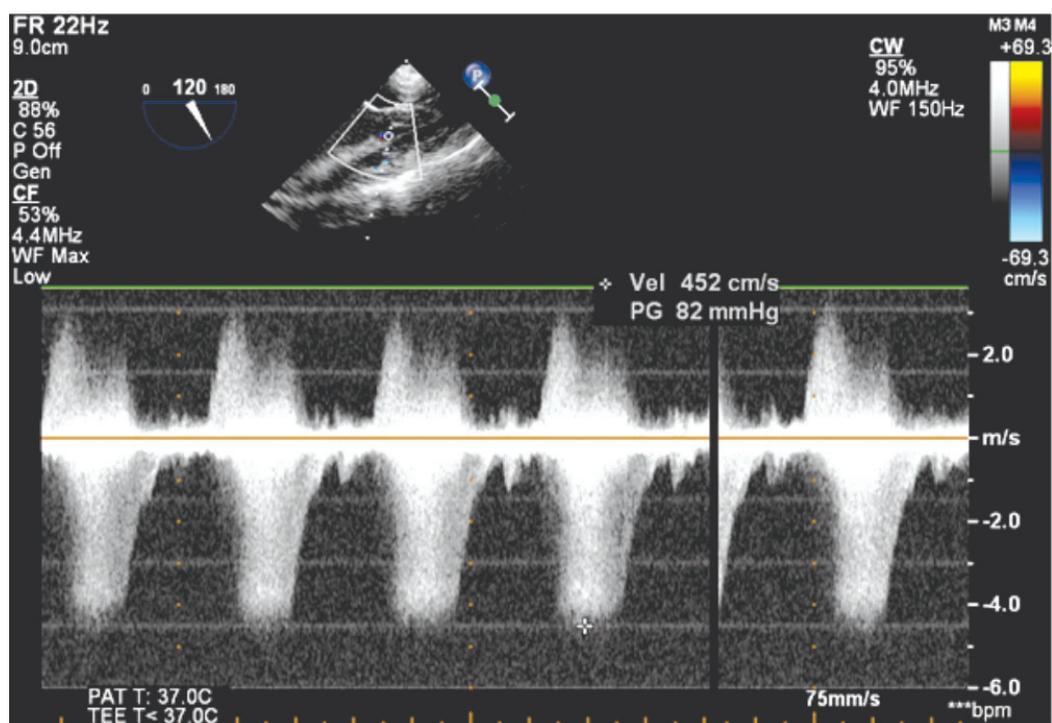


**Fig. 1.6:** Principles of continuous wave (CW) and pulsed wave (PW) Doppler. The CW transducer (A) emits and receives simultaneously through two different crystals. The PW Doppler transducer (B) emits short pulses of ultrasound waves, the distance of interrogation is determined by the reception delay; the speed of the target is calculated by the difference between emitted and received frequencies.

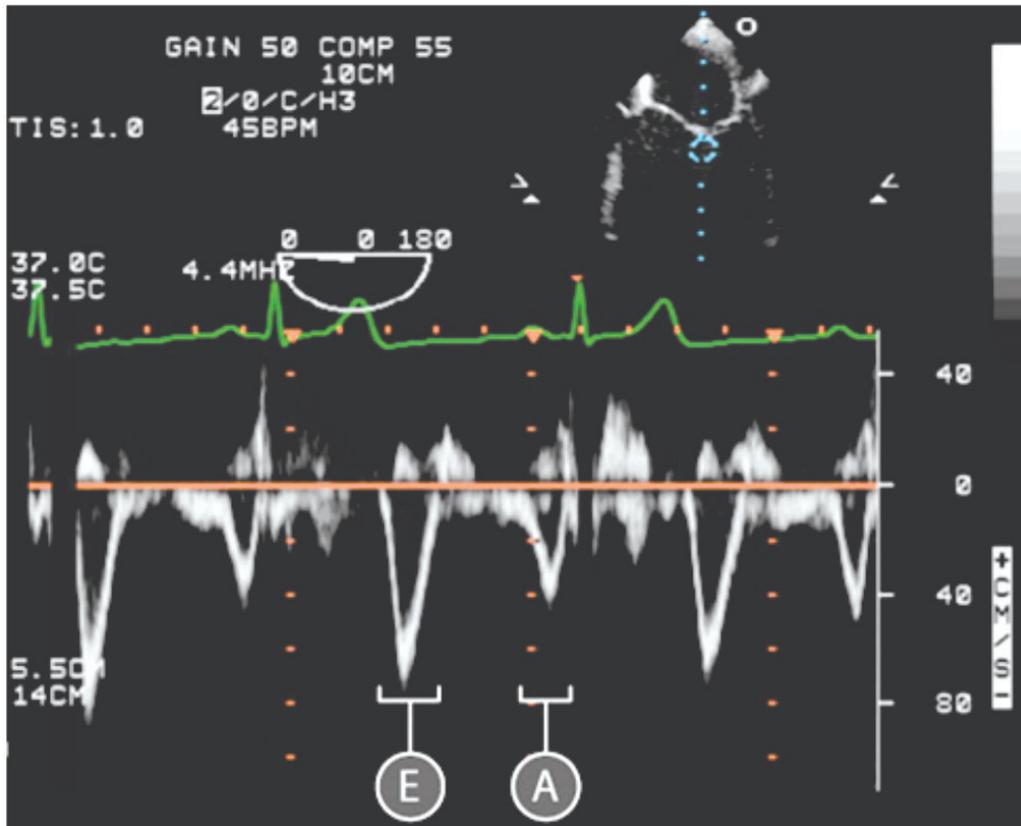
computing velocity and pressure gradients, valve orifice areas and blood flow (Figs 1.8 and 1.9). Pulsed wave Doppler information from a given field sector can be represented at multiple points at multiple depths with different colors, each color denoting a particular frequency. This is called color Doppler and forms the basis of color-flow imaging. The Doppler flow is usually superimposed on the two-dimensional cardiac image. The physiologic motion of moving flow determines the direction; traditionally red color is assigned to motion towards the transducer and blue for motion away from the transducer with multiple shades of green and yellow for intermediate velocities in a turbulent flow. Since color Doppler uses the concept of pulsed wave Doppler, higher velocities are not properly represented and result in aliasing (Fig. 1.10). As image aliases, color changes from blue to red and vice-versa. Another limitation is the higher time required for color flow imaging that reduces the frame rate. Therefore for optimizing the image resolution, the sector angle and region of interrogation for color Doppler need to be much smaller than the actual 2-D image.



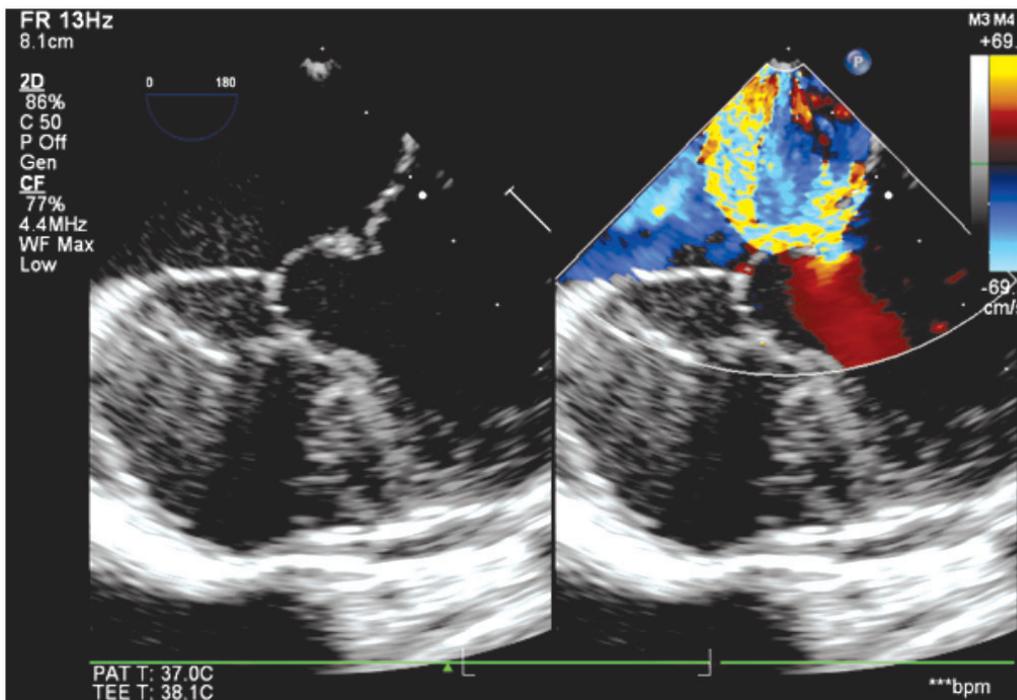
**Fig. 1.7:** A pulsed wave Doppler profile showing aliasing of the mitral regurgitation jet (arrow).



**Fig. 1.8:** A continuous wave Doppler profile depicting high velocity flow across a subaortic ventricular septal defect.



**Fig. 1.9:** A pulsed wave Doppler profile depicting the low velocity mitral inflow pattern during diastole. (E: early filling phase, A: atrial contraction).



**Fig. 1.10:** A color flow Doppler image showing aliasing of the mitral regurgitation jet. The hue changes from red towards blue across the shades of 'Vibgyor'.

# Image Optimization

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A beginner in the field of transesophageal echocardiography (TEE) often struggles to obtain a good quality image. It is important to note that unlike other imaging techniques (computerised tomography, magnetic resonance imaging), getting a good image on TEE is operator dependent. Nevertheless, some equipment (defective probe) or patient related factors (fluid/air in the stomach, emphysematous lung, calcified intracardiac structures, presence of mechanical heart valves) can influence the quality of image. The operator can learn to optimise the image by understanding the function of various knobs and buttons available on the console of the TEE machine (Fig. 2.1). In addition, the

operator should also understand the physics related to ultrasound.

The typical controls of every echocardiography machine that enable the operator to obtain a good image include the following (also refer to Chapter 1).

## TRANSDUCER FREQUENCY

Higher transmitted ultrasound frequency is chosen for interrogating shallow structures and lower frequency for deeper structures. Higher frequency ultrasound permits increased frame rate and improved temporal resolution.



Fig. 2.1: The TEE machine console showing the various knobs and controls.



### Wavelength

Longer wavelengths (lower frequency) enable greater tissue penetration, but at the cost of resolution.

### Pulse Repetition Frequency

Pulse repetition frequency is the frequency at which ultrasound pulses are generated. The resolution is improved with frequent ultrasound pulses. The velocity scale enables adjustment of the pulse repetition frequency so as to match the velocities of interest and prevent aliasing.

### Frame Rate

A lower frame rate decreases the temporal resolution, i.e., the ability to locate moving structures.

### Sector Width/Scan Line Density

The angle of the sector displayed on the monitor can vary from 15 to 90°. The temporal resolution and frame rate can be improved by reducing the sector width (increasing the line density and pulse repetition frequency) (Figs 2.2 and 2.3).

Most of the images can be considerably improved by adjustment of 3 controls, i.e. gain, depth, and time gain compensation (TGC).

### Gain

The receiver gain increases the amplitude of the returning ultrasound signals. It serves to compensate for signal loss due to attenuation. It is set at a level that permits rare noise signals into the image. A high gain setting brightens the image but also introduces false signals, whereas a low gain setting masks actual signals (Figs 2.4–2.6).

In general, the gain settings are around 75 in order to obtain a good image. However, it needs to be adjusted according to a particular image. For instance, in Fig. 2.8, the gain setting is only 14, but the image quality is quite good. Hence, there is no need to increase the gain setting here (compare with Fig. 2.7).

### Depth

Increasing the depth of interrogation increases the wavelength but decreases the frequency (frame rate) and thereby, the resolution. Conversely, the resolution is improved by decreasing the depth (Figs 2.9 and 2.10). It is ideally set to just display the structure of interest. In general, the depth setting is 8–10 cm, 10–15 cm and 15–20 cm for upper-esophageal, midesophageal and transgastric views respectively.

There are no fixed depth settings and it should be adjusted according to the area of interest. For instance, visualization of interatrial septum can be considerably improved in midesophageal view by decreasing the depth (Figs 2.11 and 2.12).

### Time Gain Compensation

Time gain compensation allows selective depth-dependent amplification. As the ultrasound reaches the far field, the returning signal becomes weak. This can be amplified by TGC. Figure 2.13A and B shows the TGC controls. It can be used to amplify weaker signals returning from far field more compared to those returning from near field (Figs 2.14 and 2.15).

### Lateral Gain Control

When the direction of ultrasound is parallel to the object, there is little reflection resulting in a weak signal. This can be augmented by lateral gain control (LGC). For instance, in transgastric midpapillary view, the lateral wall and the interventricular septum are parallel to the ultrasound waves hence, there can be a dropout of these structures. This can be improved by adjusting the LGC (Figs 2.16 and 2.17).

Lateral gain compensation allows for selective amplification along adjacent scan lines within a sector scan.

### Focus

The optimal focal depth is kept at the point of interest as the lateral resolution is maximal at this point. Sound energy travels as a beam until the focal point (near field) beyond which it diverges (far field) (Figs 2.18 and 2.19).

## COLOR CONTROLS

### Color Box Size and Position

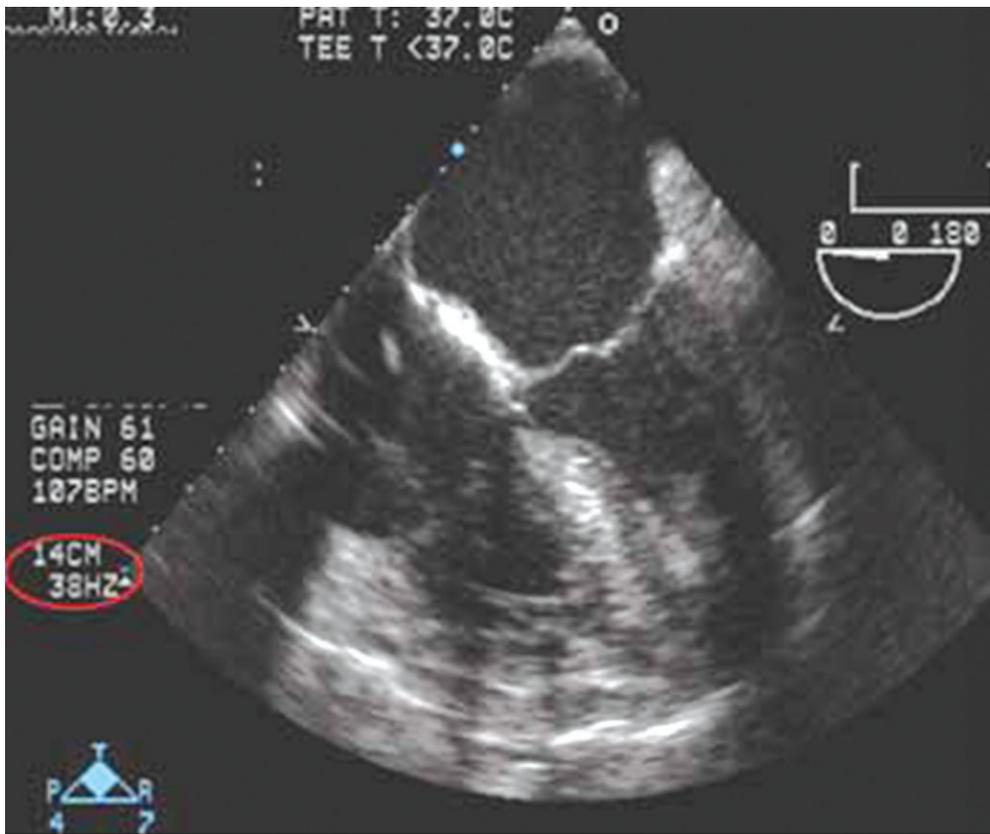
Figure 2.20 shows the TEE machine console highlighting the Doppler controls. Color is displayed in the region of interest. As the sector width or depth of color box is increased, the frame rate, and therefore, the temporal resolution, is decreased (Figs 2.21 and 2.22).

### Color Gain

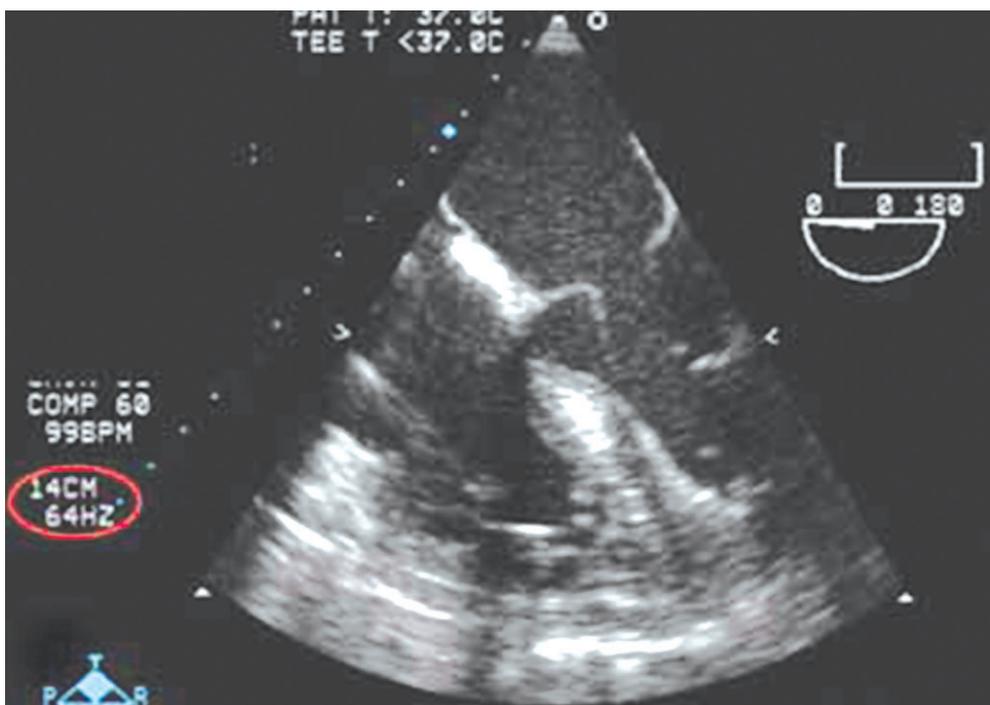
Color gain amplifies the returning ultrasound signals. It is adjusted by decreasing the color gain until speckles of color disappear outside the blood pool.

### Color Scale

The color scale must be adjusted according to the blood velocity being evaluated as aliasing occurs when velocities outside the range are sampled.



**Fig. 2.2:** Midesophageal 4-chamber view with maximum width of the sector scan.



**Fig. 2.3:** Midesophageal 4-chamber view depicting how a reduction in sector scan leads to an increase in frame rate (64 Hz) for the same depth (compare with Fig. 2.2).

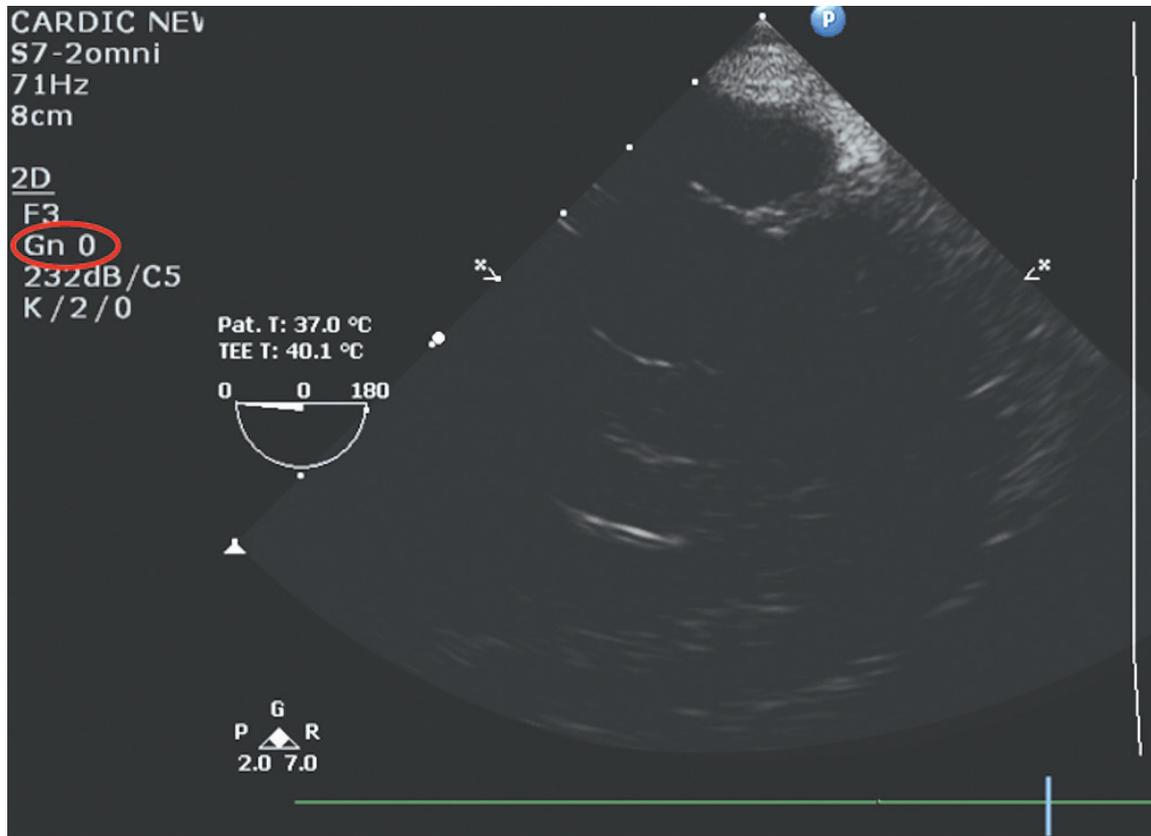


Fig. 2.4: The image is not visible when the gain is set at minimum.

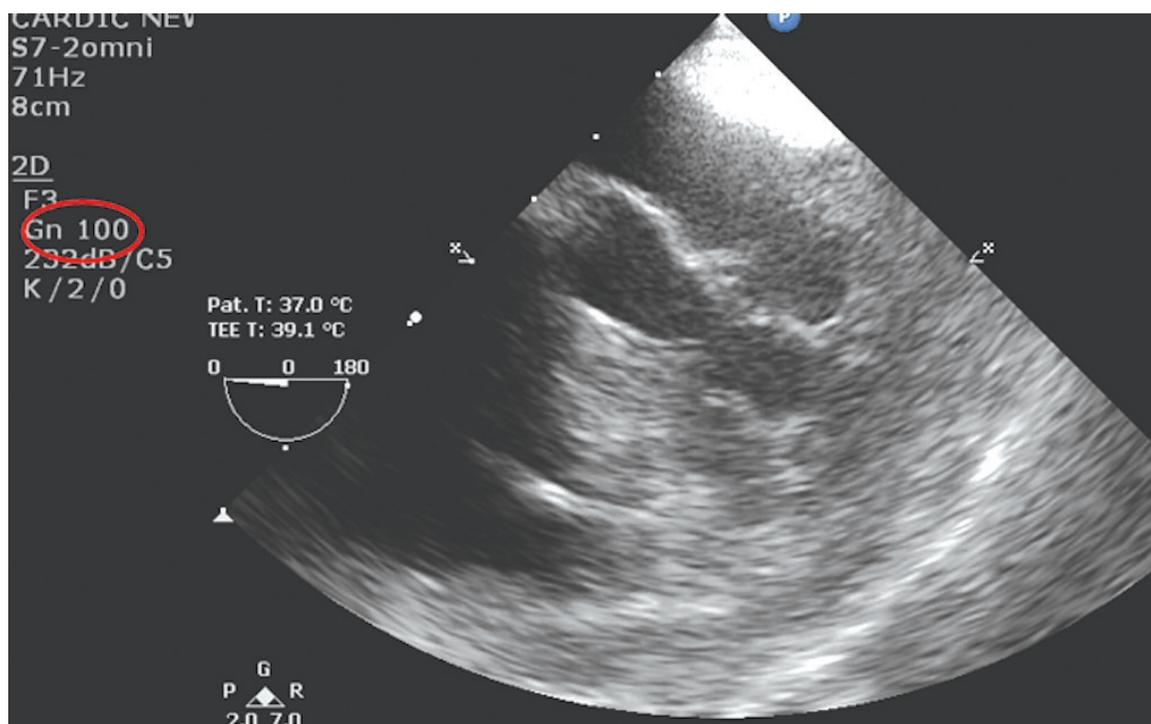


Fig. 2.5: The image appears unduly bright when the gain is set at maximum.

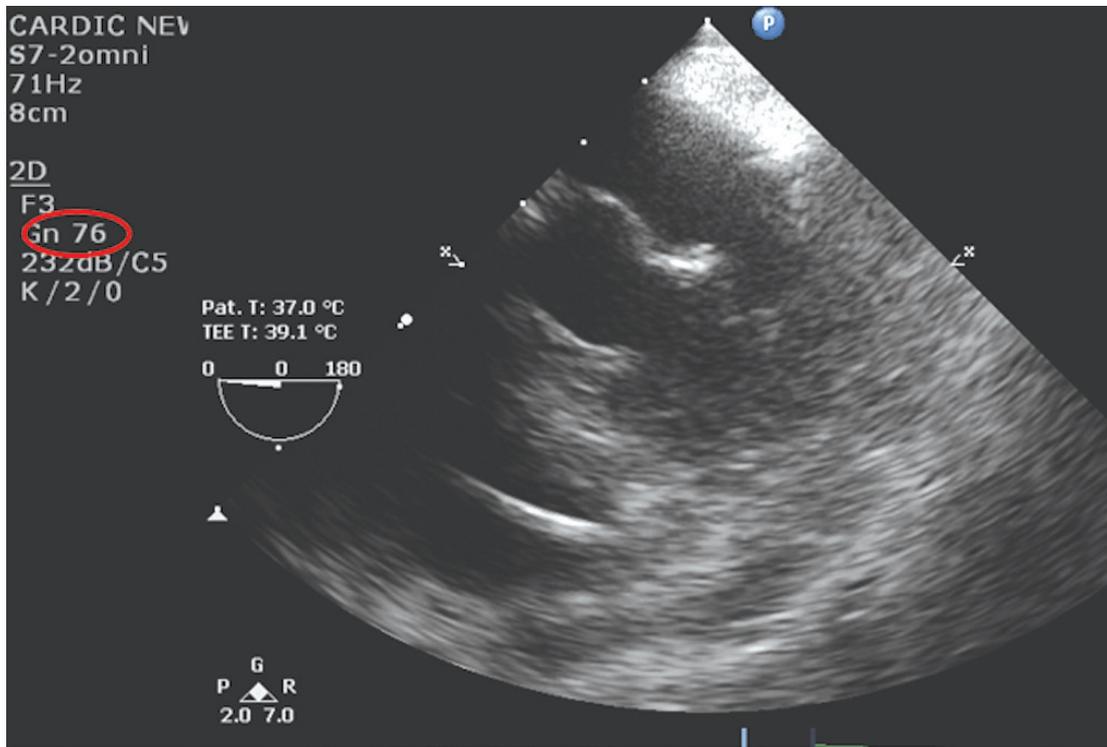


Fig. 2.6: The gain is set to permit image detailing.

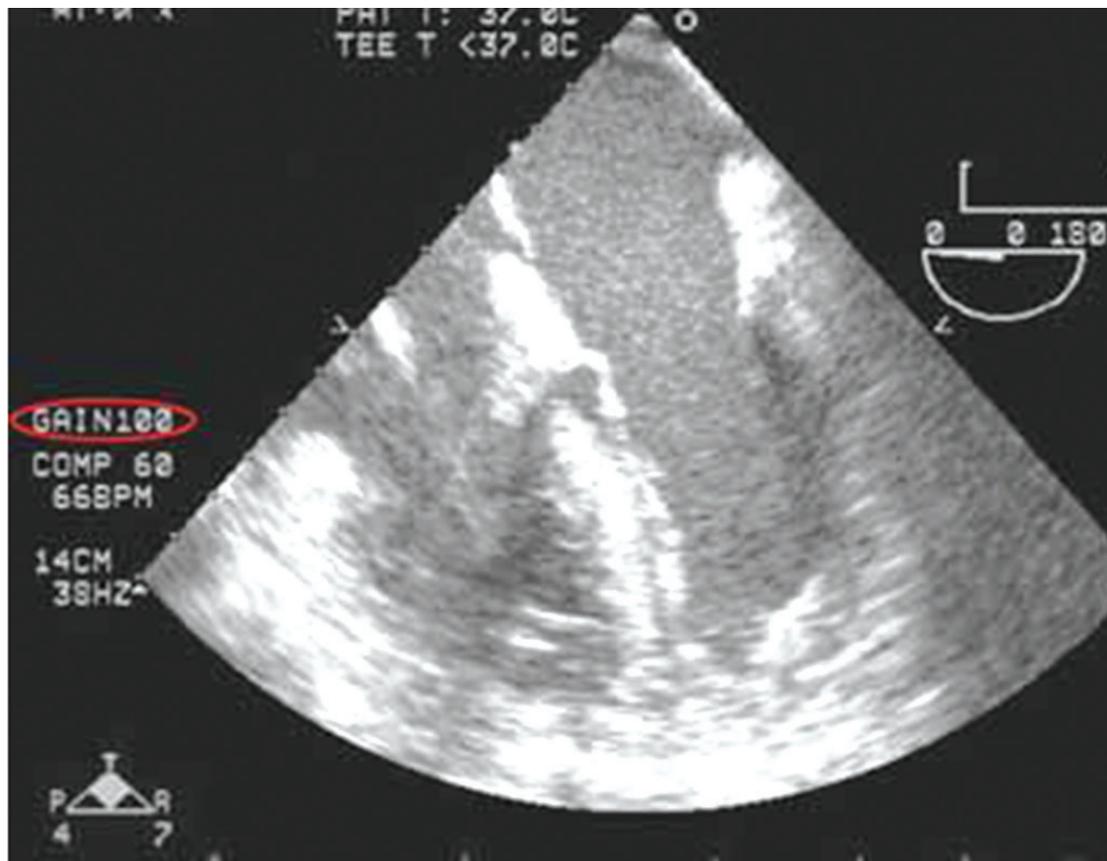
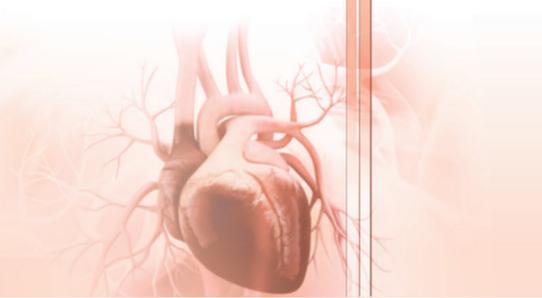
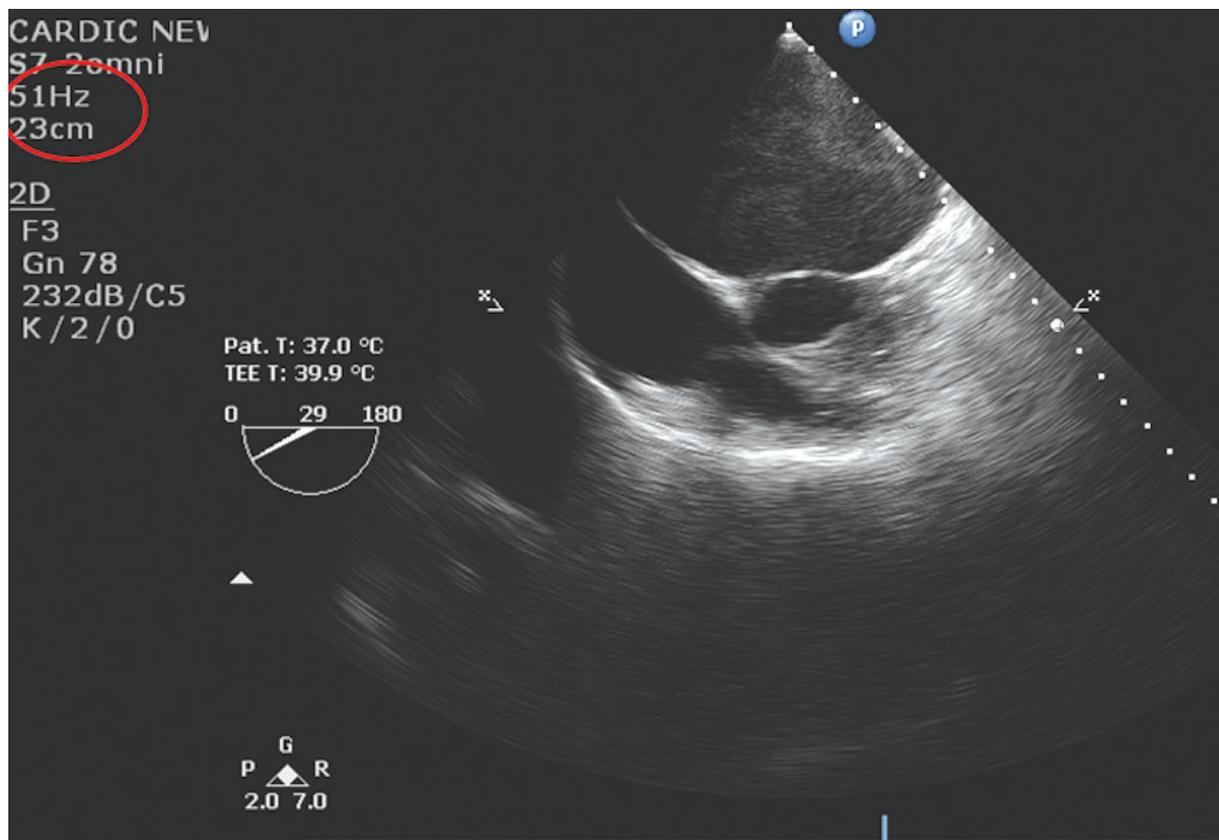


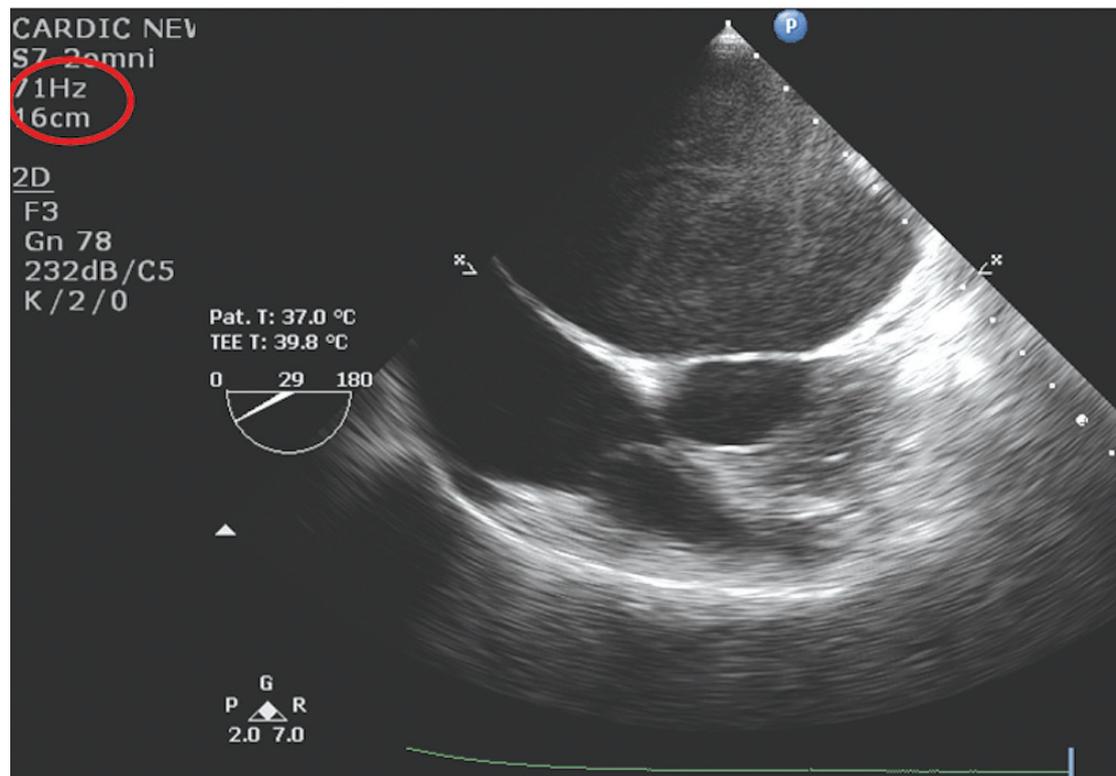
Fig. 2.7: Midesophageal 4-chamber view with increased overall gain in 2-D ultrasound.



**Fig. 2.8:** Midesophageal 4-chamber view wherein gain has been reduced to optimize the 2-D ultrasound image.



**Fig. 2.9:** Midesophageal aortic valve short-axis view showing a greater depth setting (23 cm) resulting in lower frequency and resolution. Note the frequency is 51 Hz.



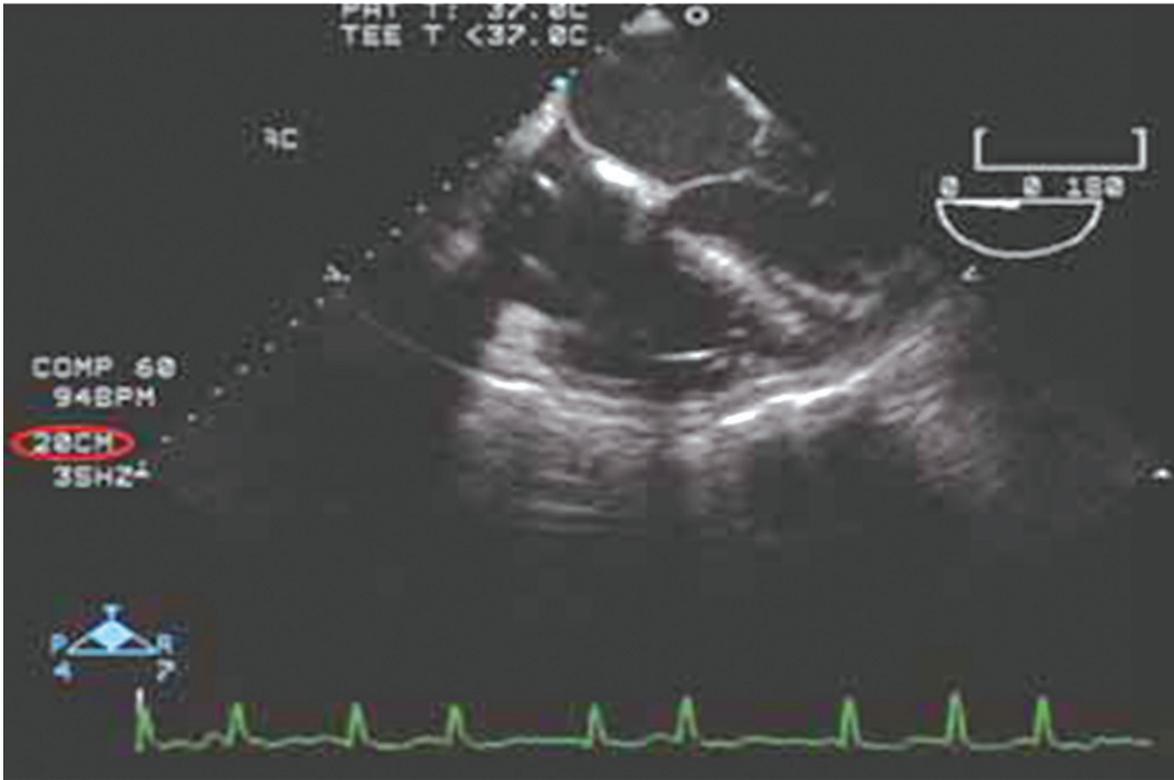
**Fig. 2.10:** Midesophageal aortic valve short-axis view showing a lower depth setting (16 cm) resulting in higher frequency and resolution. Note that the frequency has improved to 71 Hz (compare with Fig. 2.9).

### Variance

A variance in flow, i.e., turbulent flow may be color-coded (e.g.: shades of green) to distinguish it from laminar flow which is coded on the red-blue color flow map.

### Baseline

Adjustment of the baseline of the spectral display is performed to accommodate the entire velocity range in the direction of interest and prevent aliasing (Figs 2.23 and 2.24). For higher velocities, aliasing can be prevented by utilising continuous wave Doppler and increasing the scale to accommodate the velocity (Fig. 2.25).



**Fig. 2.11:** Midesophageal 4-chamber view showing the depth set at 20 cm and the resultant frequency 35 Hz resulting in compromised resolution.



**Fig. 2.12:** Midesophageal 4-chamber view wherein the depth of imaging has been decreased to 9 cm resulting in increased frequency and improved resolution. In this case, it is useful for examination of inter-atrial septum.

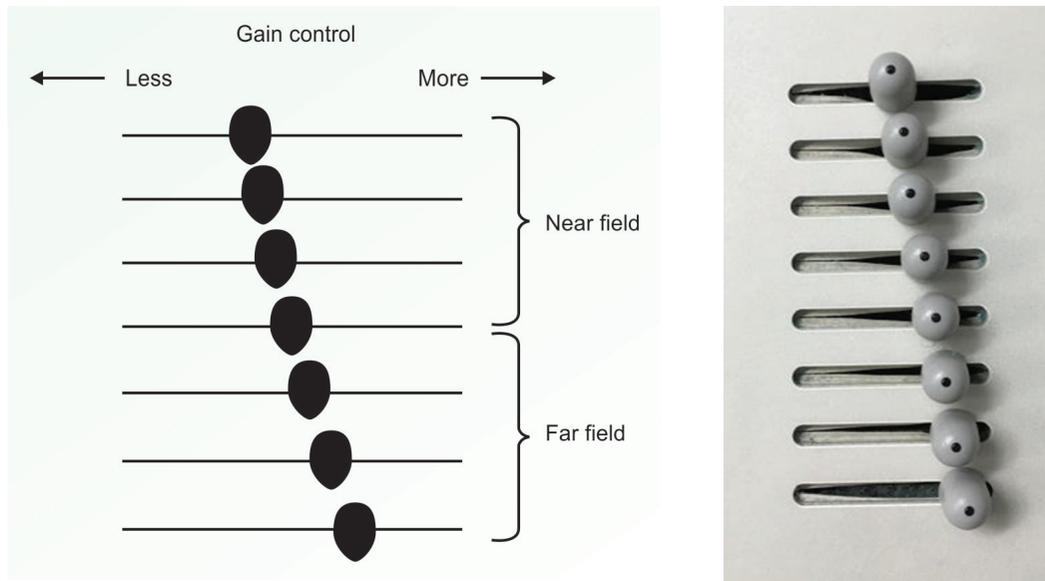


Fig. 2.13: The time gain compensation controls on the TEE machine console.

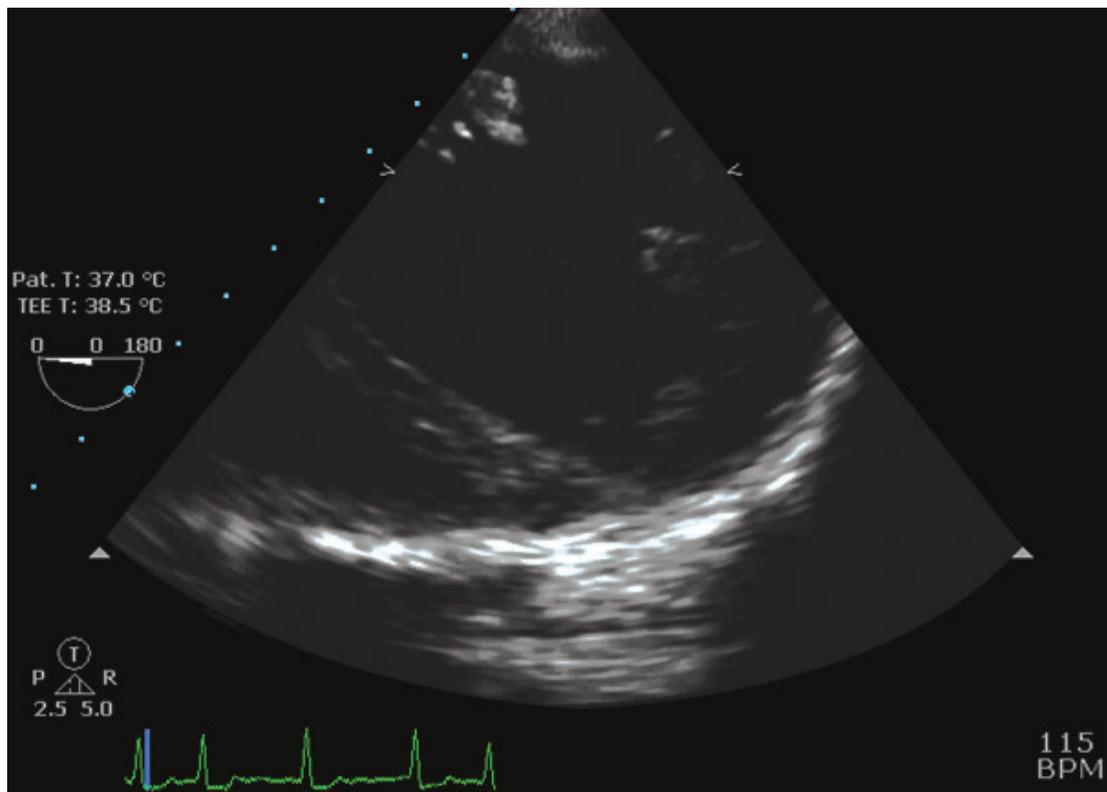
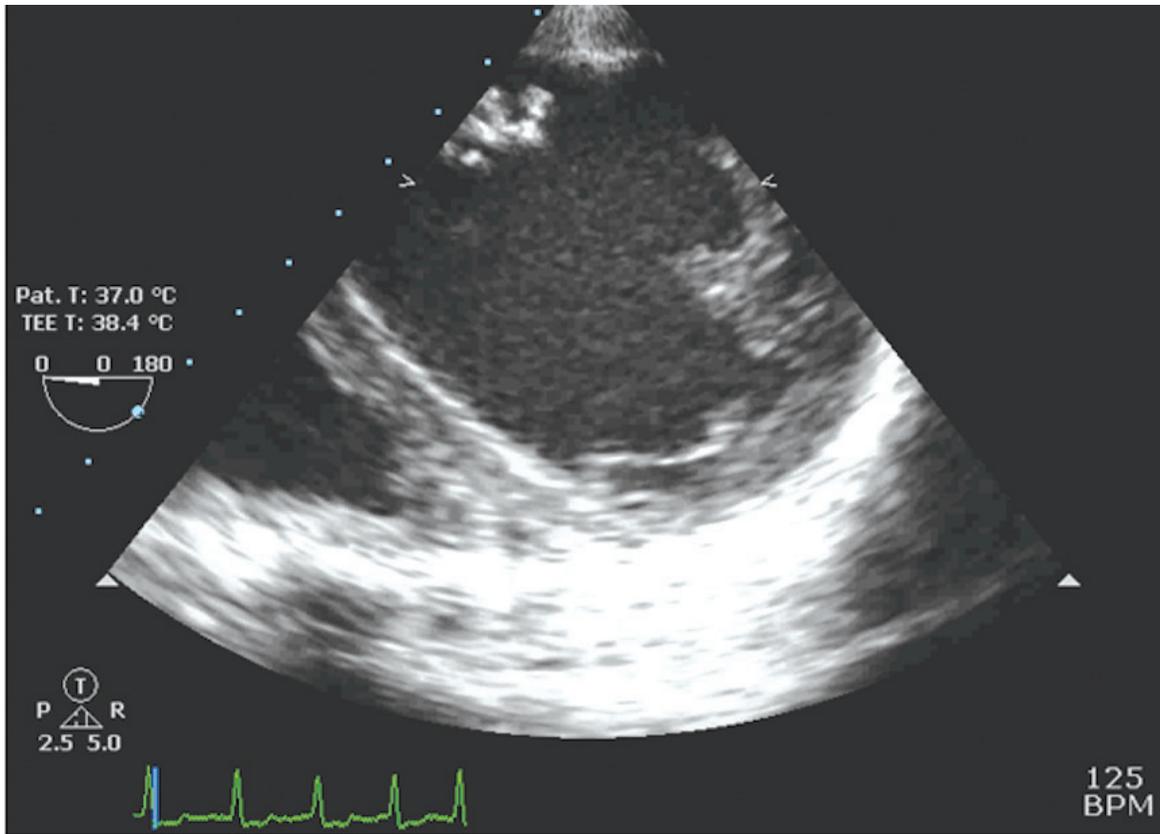
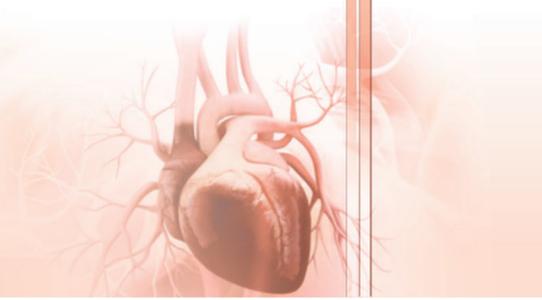
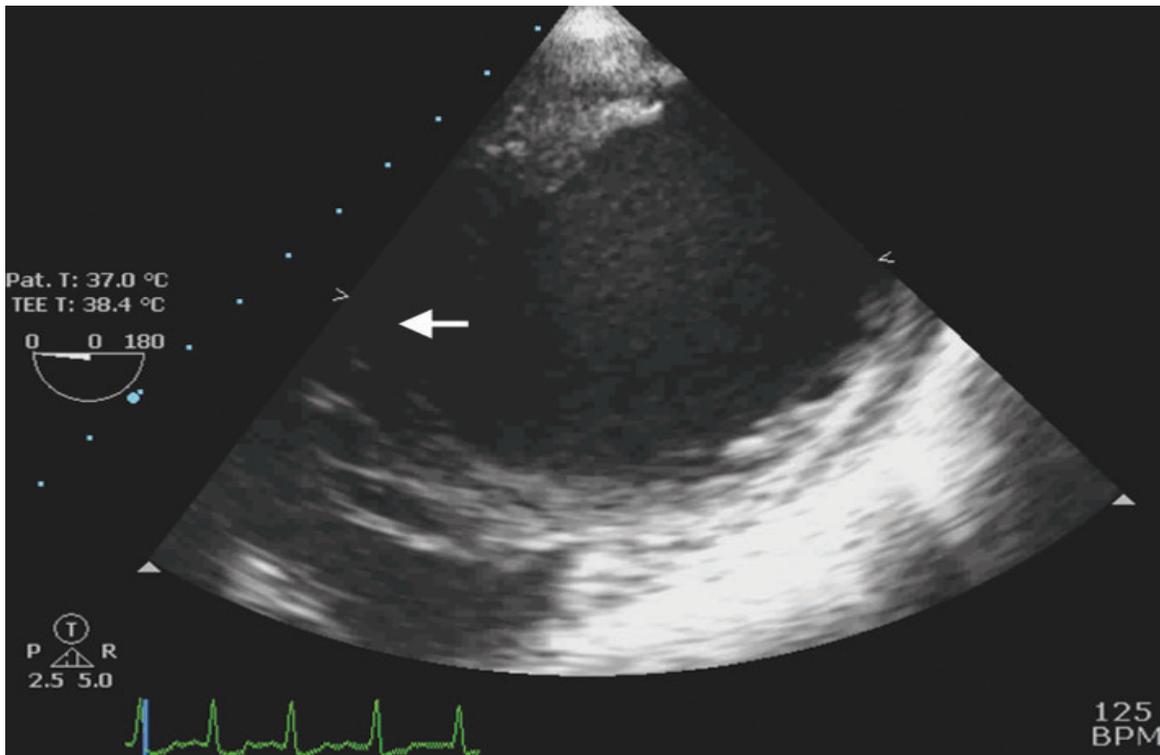


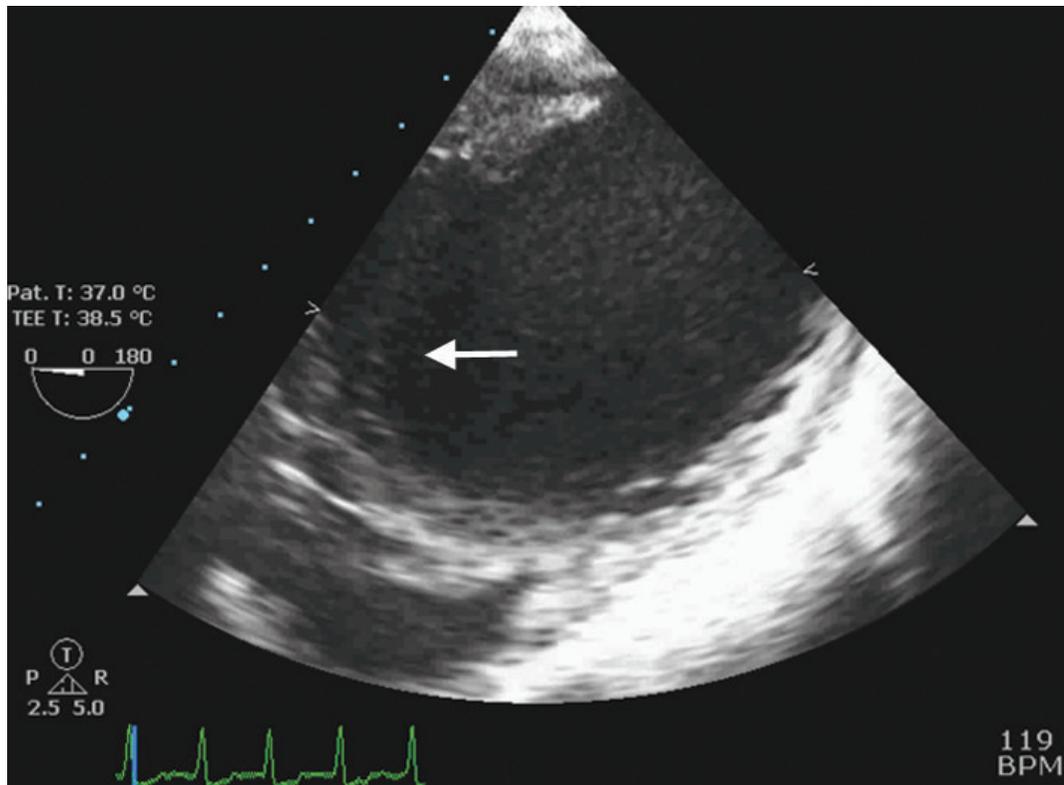
Fig. 2.14: The time gain compensation setting is low in this figure, the image appears less bright.



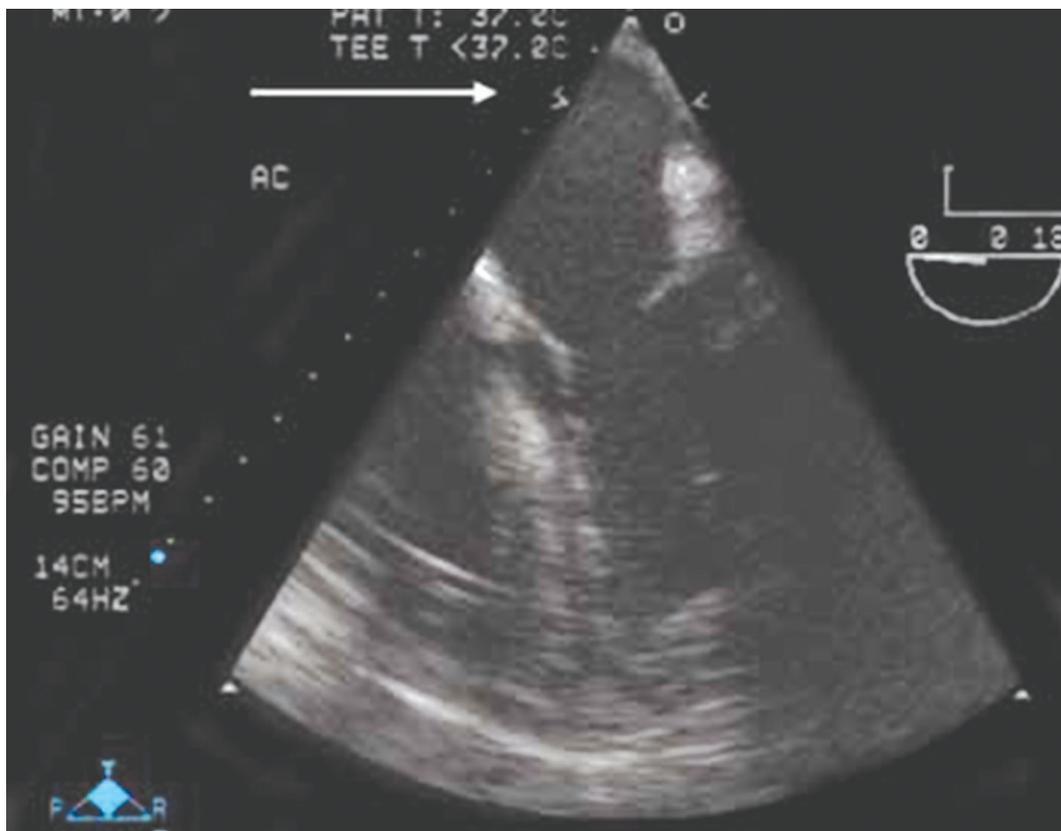
**Fig. 2.15:** The image is brighter at the bottom as the time gain compensation has been increased in the far field.



**Fig. 2.16:** Transgastric mid-papillary short-axis view. Note the image drop out of the interventricular septum as the ultrasound wave strikes parallel to the septum (arrow).



**Fig. 2.17:** The same image as in Fig 2.16. Note the improved visualization of the interventricular septum by adjusting the lateral gain control.



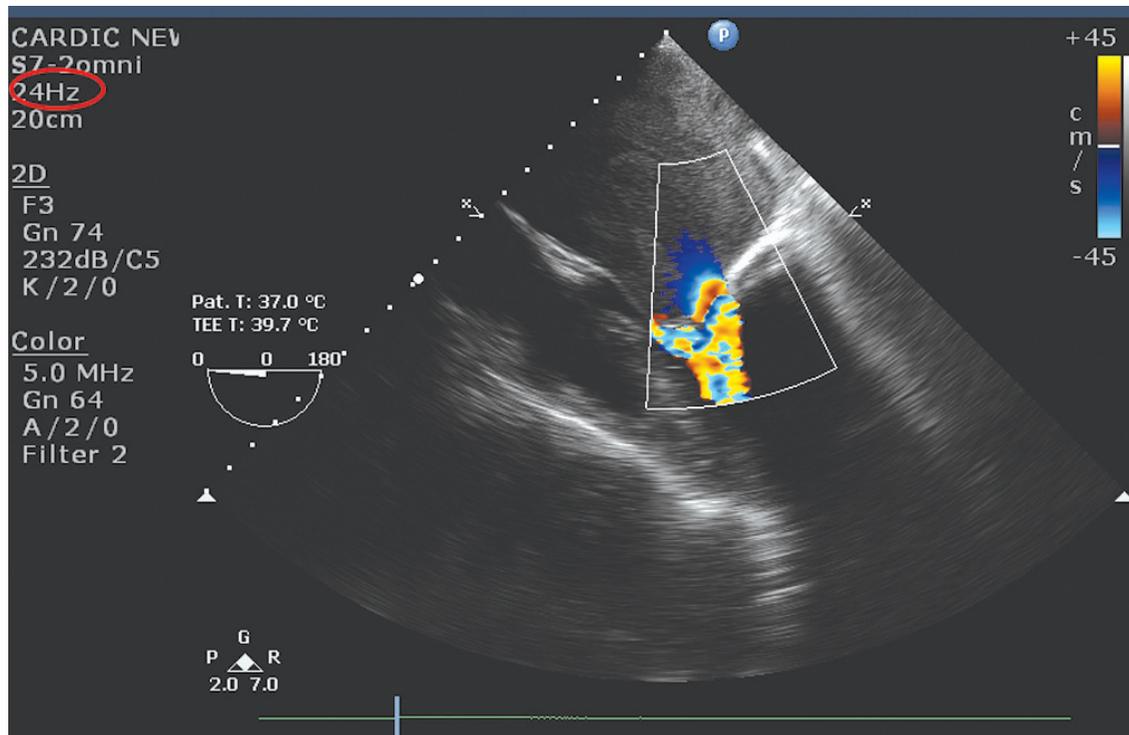
**Fig. 2.18:** With the focus set close to the probe (arrow), the far zone increases and resolution is poor beyond the focal point.



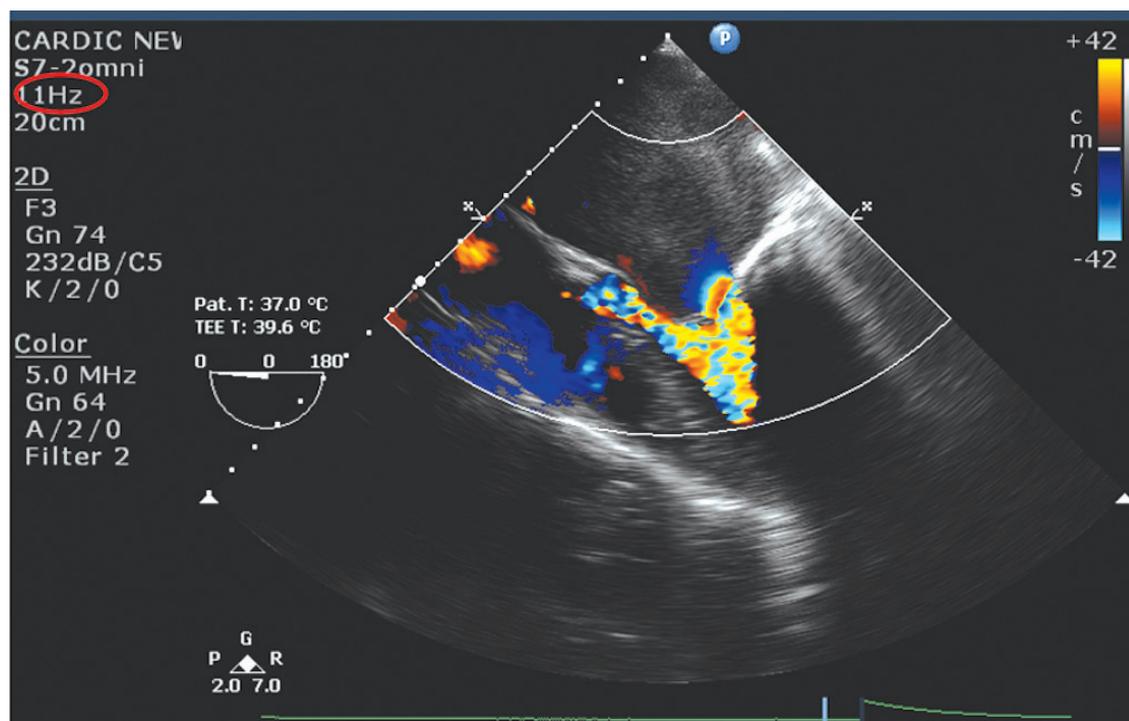
Fig. 2.19: The focus is now set at the point of interest (valvular level) resulting in improvement of image quality.



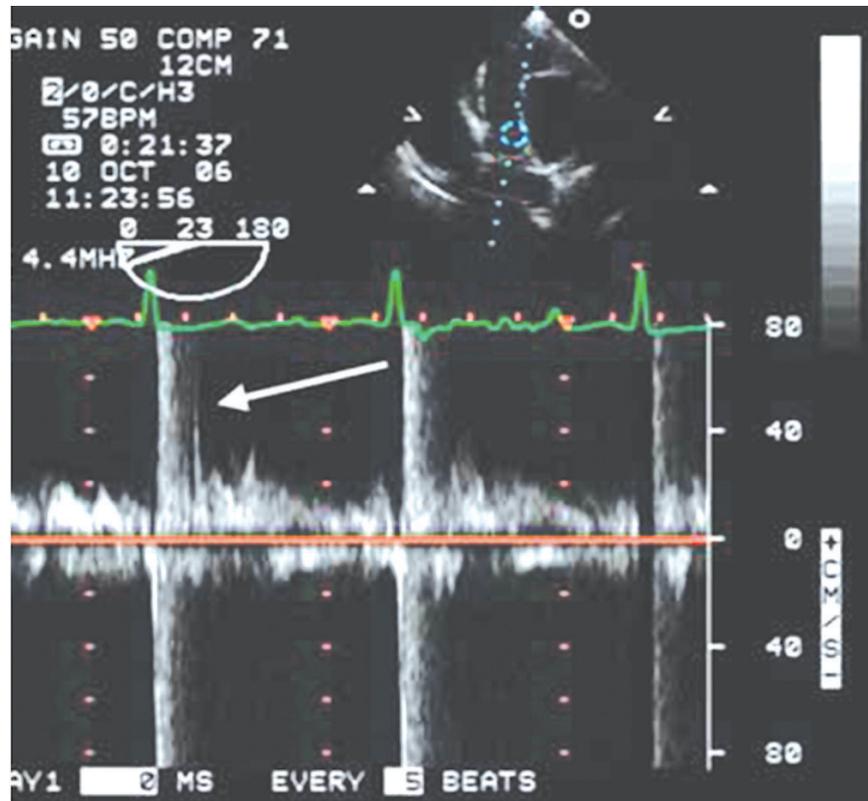
Fig. 2.20: The TEE machine console highlighting the color Doppler controls.



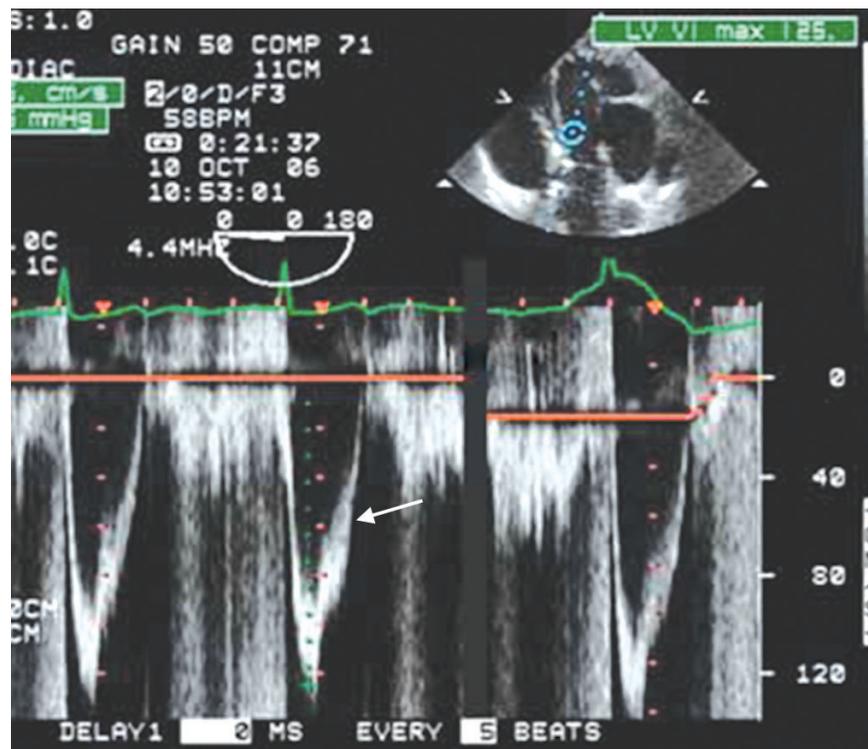
**Fig. 2.21:** Midesophageal 4-chamber view showing that a narrow color box increases the frequency (24 Hz) and thereby, the resolution.



**Fig. 2.22:** Midesophageal 4-chamber view showing that a wider color box decreases the frequency (11 Hz) and thereby, the resolution (compare with Fig. 2.21).



**Fig. 2.23:** Deep transgastric view showing cut-off of the aortic flow profile and its appearance above the baseline (arrow), implying aliasing.



**Fig. 2.24:** Shifting the baseline upwards accommodates the entire pulse wave Doppler envelope (arrow).