Textbook of Echocardiography for Intensivists and Emergency Physicians

Armando Sarti
F. Luca Lorini
Editors

Second Edition

Springer

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Preface

The first edition of this book, *Echocardiography for Intensivists*, was published six years ago. At that time, the use of ultrasound machines was not yet a routine for many intensivists around the world. Today, many doctors, who are involved in the management of unstable or critically ill patients, are simply unable to practice without it. Echocardiography is now firmly considered to be the most useful diagnostic tool for the evaluation of patients with acute cardiovascular disorders in the ICU, the operating theatre and the emergency department. Ultrasound examination of the heart and great vessels provides reliable functional anatomy and hemodynamic assessment of the cardiovascular system. There is no alternative way to assist and guide the clinicians so that, within just a few minutes, they may understand the cause of cardiocirculatory shock and accordingly treat hemodynamic derangement at the bedside.

Today, more than 60 years after its initial conception, echocardiography is established enough to be in a widespread use even outside the cardiology unit. Ultrasound assessment of the cardiovascular system is continuously being refined according to advances in research and technology. However, the appropriate clinical application of this unique diagnostic technique in the acute setting requires specific education with constant updates and training.

The first edition of the book was developed from awareness of the need for a text specifically written for intensivists, emergency physicians and anaesthesiologist who wished to incorporate the ultrasound technique into their clinical practice. Thanks to the contributions made by all the authors, the first edition of the book has been well-received all around the world, and a Chinese language edition has also been published. This second edition has been completely revised and expanded with the aim of becoming a *Textbook of Echocardiography for Intensivist and Emergency Physicians*. Each existing chapter has been updated, with the addition of many new figures and references, and ten entirely new chapters have been added.

We wish to express our gratitude to all of the authors of both editions for their invaluable contributions, and we would also like to thank Springer for their skilful production of this textbook. Last, but not least, we are indebted to our families for their understanding and support.

Finally, we wish to dedicate our efforts to all the patients who have to face critical medical conditions and to the doctors, nurses and caregivers who do their best to provide them with the care they need.

We will feel greatly rewarded if this book helps to improve, in some way, the care of such patients.

Firenze, Italy Armando Sarti
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Part IV Echocardiography in the ICU and OR: Basic and Advanced Applications

Part I

Ultrasound and Use of the Echo Machine

1 Essential Physics of Ultrasound and Use of the Ultrasound Machine

Dionisio F. Colella, Paolo Prati, and Armando Sarti

1.1 Ultrasound

Sound is a mechanical wave made up of compressions and rarefactions of molecules in a medium (solid, liquid, or gas) (Fig. 1.1).

Sounds is characterized by some parameters:

- *Frequency* is the number of cycle per unit time (1 s), measured in hertz (Hz). The higher the frequency, the better the resolution, but the lower the penetration (Fig. 1.2).
- *Period* is the duration of a cycle (the inverse of frequency).
- *Wavelength* is the distance that sound travels in one cycle. The wavelength depends on the size of the piezoelectric crystals in the transducer and the medium through which the sound wave travels (Table 1.1).
- *Amplitude* is the amount of change in the oscillating variable. Amplitude decreases as the wave travels (attenuation), leading to echoes from deeper structures being weaker than those from superficial structures. It is measured in decibels:

$$\text{Decibel}(\text{dB}) = 20\log_{10} A^2 / A_r^2,$$

where A is the sound amplitude of interest and A_r is a standard reference sound level.

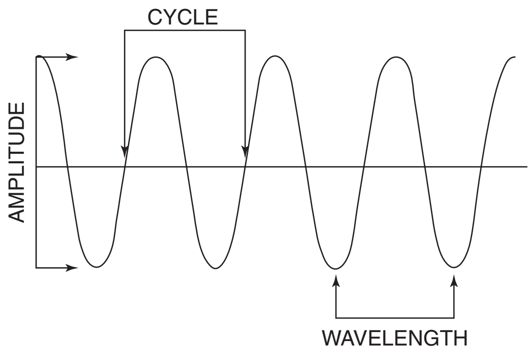

Fig. 1.1 A sound wave

- *Intensity* is the measure of the energy in a sound beam. It is related to potential tissue damage. For example, ultrasound used for lithotripsy has high intensity to fragment renal stones. It is measured in watts per square meter.
- *Power* is the amount of energy transferred. It is expressed in watts.
- The power or the intensity levels are not represented on the ultrasound machine, but there are two other variables that indirectly change those two parameters: mechanical index and

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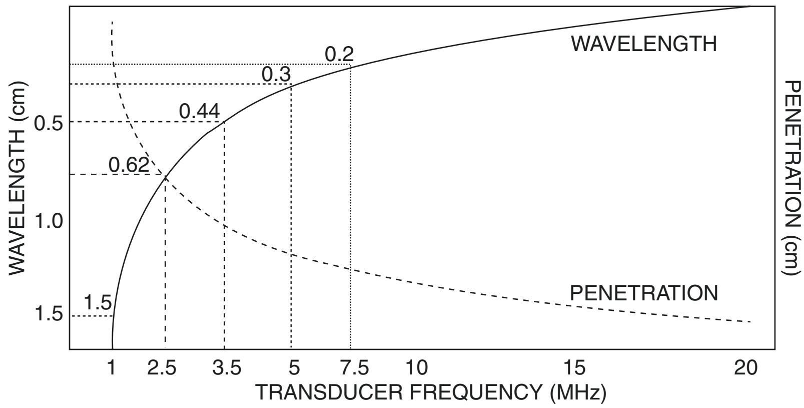

Fig. 1.2 Relationship between transducer frequency, penetration, and wavelength. As the transducer frequency increases, resolution increases and penetration decreases

Table 1.1 Relationship between frequency and wavelength

Frequency (MHz)	Wavelength (mm)
1.25	1.2
2.5	0.60
5.0	0.30
7.5	0.20
10.0	0.15

Table 1.2 Ultrasound velocities in different mediums

Material	Velocity (m/s)
Air	330
Water	1497
Fat	1440
Blood	1570
Soft tissue	1540

thermal index. The first one represents the risk of cavitation. The second one is related to the increase of temperature of the tissues (Table 1.1, Fig. 1.2).

- *Propagation velocity* is the velocity determined by the medium that the sound passes through. It is related to the tissue's resistance to compression. Velocity is the product of frequency and wavelength. The propagation velocity through a medium is increased by increasing stiffness of the medium and is reduced by increasing density of the medium (Table 1.2).

Velocity is the product of wavelength and frequency:

$$v = \lambda \times f.$$

1.2 Interaction of Ultrasound with Tissues

1.2.1 Attenuation

When the ultrasound beam passes through uniform tissues, its energy is attenuated by dispersion and absorption.

Absorption is the conversion of ultrasound energy into heat. The attenuation coefficient relates the amount of attenuation to the frequency of the ultrasound beam and the distance that beam travels.

Dispersion occurs because of reflection, refraction, and scattering. The attenuation of the sound wave is increased at higher frequencies, so in order to have better penetration of deeper tissues, a lower frequency is used.

Attenuation involves less energy returning to the transducer, resulting in a poor image.

As the sound traveling through a tissue reaches another tissue with different acoustic properties, the sound energy can be reflected or change its direction, depending on the acoustic impedance of the second interface.

Acoustic impedance is the ability of a tissue to transmit sound and depends on:

- The density of the medium
- The propagation velocity of ultrasound through the medium:

$$Z = \rho \times v,$$

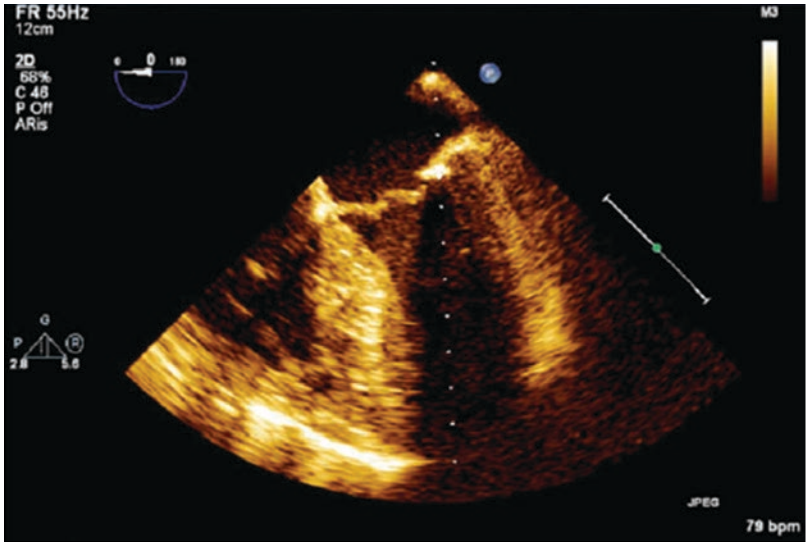

Fig. 1.3 The left ventricular wall is hidden behind a calcified posterior mitral leaflet

where Z is the acoustic impedance, ρ is the density of the material, and v is the speed of ultrasound.

If different mediums have a large difference in acoustic impedance, there is an acoustic impedance mismatch. The greater the acoustic mismatch, the greater the percentage of ultrasound reflected and the lower the percentage transmitted.

1.2.2 Reflection

When a sound wave reaches a smooth surface, it is reflected with an angle that is opposite the incident angle. *The more the angle is near 90°, the lower the amount of energy that is lost.*

There are two types of *reflection*:

1. *Specular* reflection
2. *Scattering* reflection

If the sound wave reaches a small and irregularly shaped surface (such as red blood cells), the ultrasound energy is scattered in all directions.

Reflection can be measured by the reflection coefficient:

$$R = \left(Z_2 - Z_1\right)^2 / \left(Z_2 + Z_1\right)^2,$$

where R is the reflection coefficient and Z is the acoustic impedance.

When the second medium encountered is a strong reflector, some phenomena can occur:

- Acoustic shadowing (Fig. 1.3)
- Reverberation (Fig. 1.4)
- A side lobe (Fig. 1.5)

1.2.3 Refraction

When a sound beam reaches the interface between two mediums, some of it is not reflected but passes through the interface, and its direction is altered. This is called refraction. The amount of refraction is proportional to the difference in the velocity of sound in the two tissues and to the angle of incidence:

$$n_1 / n_2 = \sin\theta_1 / \sin\theta_2,$$

where n is the refraction coefficient and θ is the angle of incidence.

It is possible to see some refraction artifact (Figs. 1.6 and 1.7).

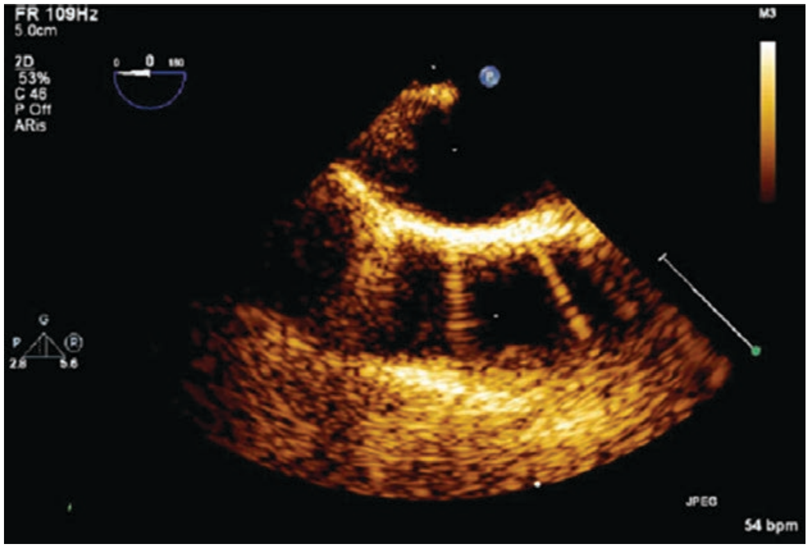

Fig. 1.4 Comet tail. Mirror image: double-barred aorta

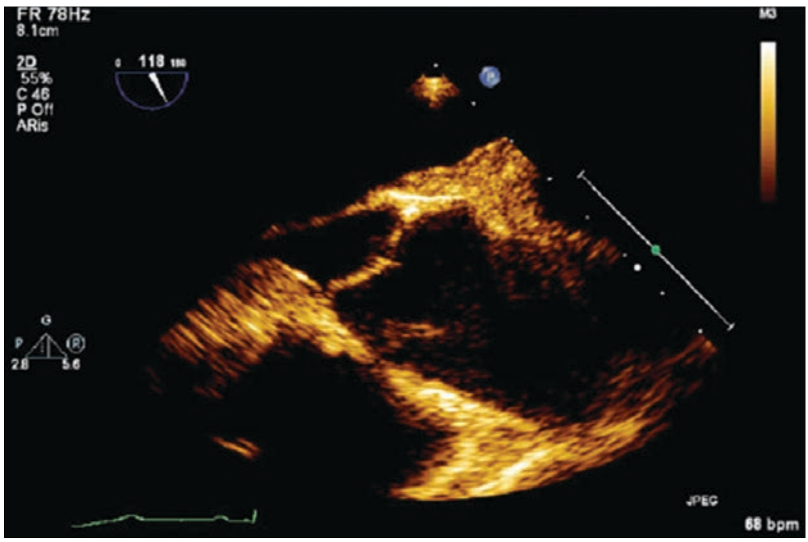

Fig. 1.5 Side lobe artifacts can create a false aortic flap

1.3 Ultrasound Wave Formation

Ultrasound waves are generated by *piezoelectric crystals*. An electrical current applied to a crystal causes vibration and consequent expansion and contraction. These changes are transmitted into the body as ultrasound waves. Modern transducers are both transmitters and receivers.

There is a strict relationship between time, distance, and velocity of ultrasound propagation. Knowing the time required for sound to travel from the transducer to an object, the time needed for the returning echo from that object to the transducer, and the propagation velocity in that medium allows one to calculate the distance the ultrasound waves have crossed. This is the basis of ultrasonic imaging.

Electrical energy is not applied to the transducer in a continuous way: ultrasound waves are produced at regular intervals with a pulsed

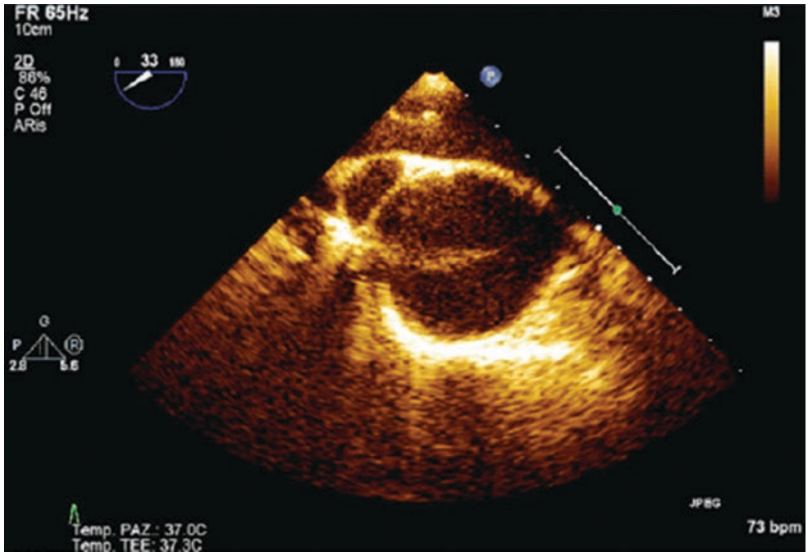

Fig. 1.6 Grading lobe. The pulmonary catheter seems to be in the aorta

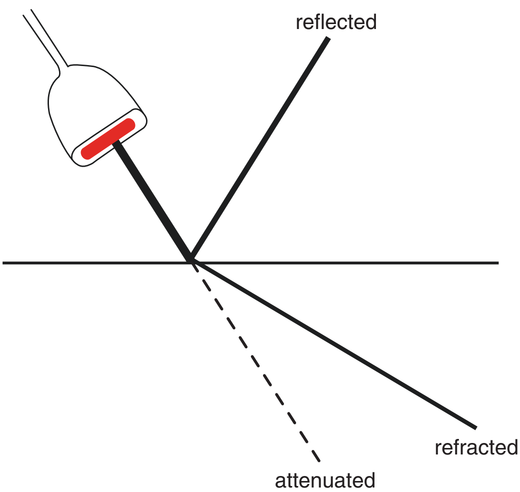

Fig. 1.7 Reflection, refraction, and attenuation

repetition period, leading to a defined *pulse repetition frequency* (PRF; in kilohertz). The wavelength of the ultrasound generated is inversely related to the thickness of the piezoelectric elements.

The piezoelectric elements cannot emit a second pulse until the first has returned to the transducer: the ability to recognize different objects is related to the frequency of emission of the ultrasound wave pulse.

The ultrasound beam emitted from the transducer has a particular shape: it begins with a narrow beam (near field), and then the ultrasound beam diverges in the far field. The length of the near field (or Fresnel zone) is related to the diameter of the transducer (D) and the wavelength:

$$L_n = D^2 / 4\lambda.$$

Even the angle of divergence, forming the far field (or Fraunhofer zone), is related to the diameter of the transducer (D) and the wavelength:

$$\sin\theta = 1.22\lambda / D.$$

The resolution is improved in the near field because of the narrower diameter of the ultrasound beam. It is easy to understand that a high-diameter transducer with high frequency (short wavelength) can produce the best ultrasound beam.

There is another way to reduce the diameter of the ultrasound beam and thus improve the resolution: focusing the beam. This produces a reduction of the beam size at a particular point, ameliorating the image.

1.3.1 Resolution

This is the ability to recognize two objects. Spatial resolution is the ability to differentiate two separate objects that are close together. Temporal resolution is the ability to place structures at a particular time.

1.3.2 Axial Resolution

This is the ability to recognize two different objects at different depths from the transducer along the axis of the ultrasound beam (Figs. 1.8 and 1.9):

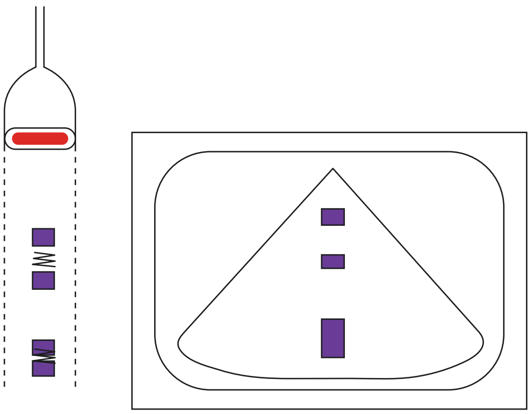

Fig. 1.8 Axial resolution. The spatial pulse length is short enough to be placed within two different structures, so they are resolved

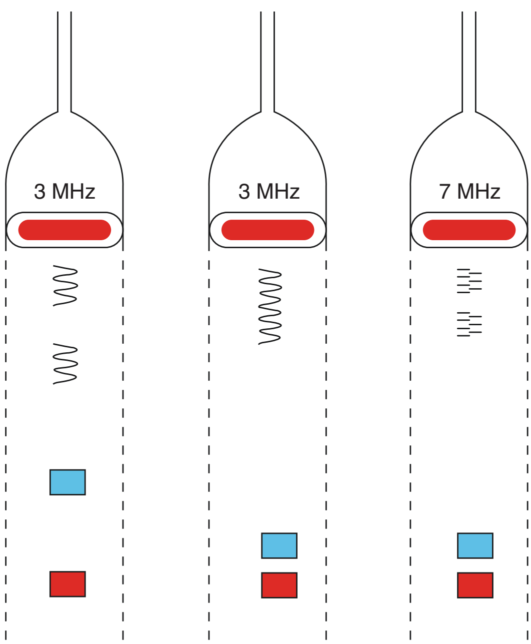

Fig. 1.9 Axial resolution and transducer frequency. Closer objects cannot be resolved by a low transducer frequency. Increasing the transducer frequency (shortening the spatial pulse length and duration) is required to resolve the objects

$$\text{Axial resolution} = \frac{\text{spatial pulse length (SPL)}}{2},$$

where SPL = $\lambda \times$ no. of cycles.

It is improved by higher-frequency (shorter-wavelength) transducers but at the expense of penetration. Higher frequencies are therefore used to image structures close to the transducer.

1.3.3 Lateral Resolution

This is the ability to distinguish objects that are side by side. It is dependent on the beam width because two objects side by side cannot be distinguished if they are separated by less than the beam width. It is improved by the use of higher-frequency transducer (which increases the beam width) and an optimized focal zone (Figs. 1.10 and 1.11).

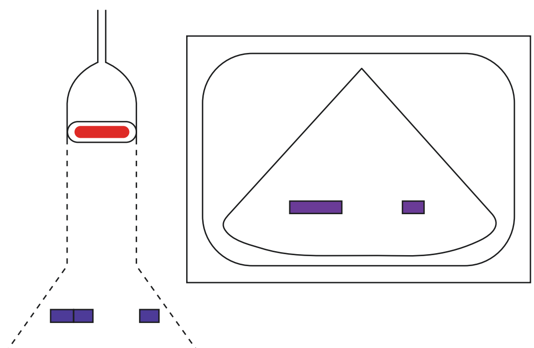

Fig. 1.10 Lateral resolution. Wider beams cannot resolve near objects

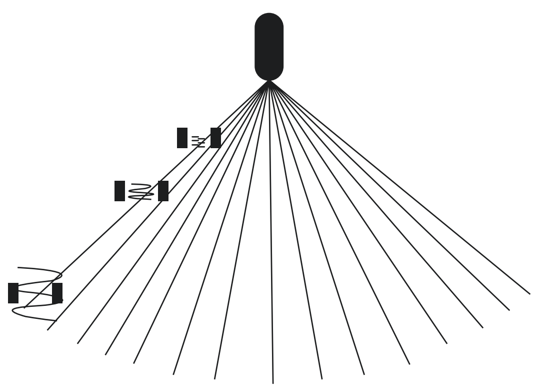

Fig. 1.11 Lateral resolution. At low depth, lateral resolution is worsened

1.3.4 Temporal Resolution

This is dependent on the frame rate. It is improved by:

- Minimizing depth—the maximum distance from the transducer as this affects the PRF
- Narrowing the sector to the area of interest—narrowing the sector angle
- Minimizing the line density (but at the expense of lateral resolution)

1.4 Doppler Echocardiography

Doppler echocardiography is a method for detecting the direction and velocity of moving blood within the heart.

The *Doppler effect* (or *Doppler shift*) is the change in frequency of a wave for an observer moving relative to the source of the wave (Fig. 1.12).

When the source of the sound wave is moving toward the observer, each successive wave is emitted from a position closer to the observer than the previous wave, and it takes less time than the previous wave to reach the observer. Then the time between the arrival of successive waves is reduced, resulting in a higher frequency. If the source of waves is moving away from the observer, the opposite effect can be seen, with increased time between the arrival of successive waves, giving them a lower frequency.

The amount of that change in frequency is the Doppler shift. Blood flow velocity (V) is related to the Doppler shift by the speed of sound in blood (C) and the intercept angle (θ) between the ultrasound beam and the direction of blood flow:

$$\text{Doppler shift} = 2 \times F(\text{transmitted}) \times \left[(V \times \cos\theta) \right] / C$$

A factor of 2 is used because the sound wave has a "round-trip" transit time to and from the transducer. If the ultrasound beam is not parallel to blood flow, an angle of incidence greater than 30° can underestimate the Doppler shift.

There are two kinds of Doppler application: pulsed-wave Doppler and continuous-wave Doppler.

In the *continuous-wave Doppler* technique, the transducer continuously transmits and receives ultrasound waves (Fig. 1.13).

The continuous-wave Doppler technique measures all velocities along the ultrasound beam. It cannot discriminate the time interval from the emission and the reflection, giving no information about the depth of the received signal. The continuous-wave Doppler technique is able to detect very high velocities, and it can be useful to evaluate the high velocity flow through a stenotic aortic valve.

In the *pulsed-wave Doppler* technique, the transducer alternately transmits and receives the ultrasound wave and its returning echo (Fig. 1.14).

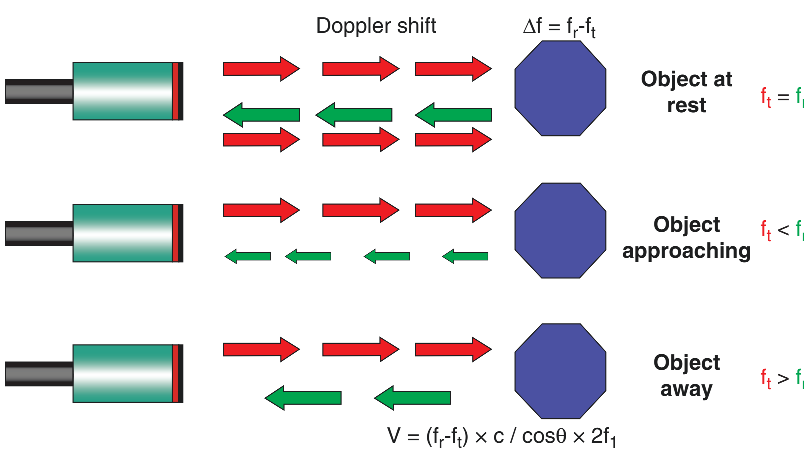

Fig. 1.12 Doppler shift

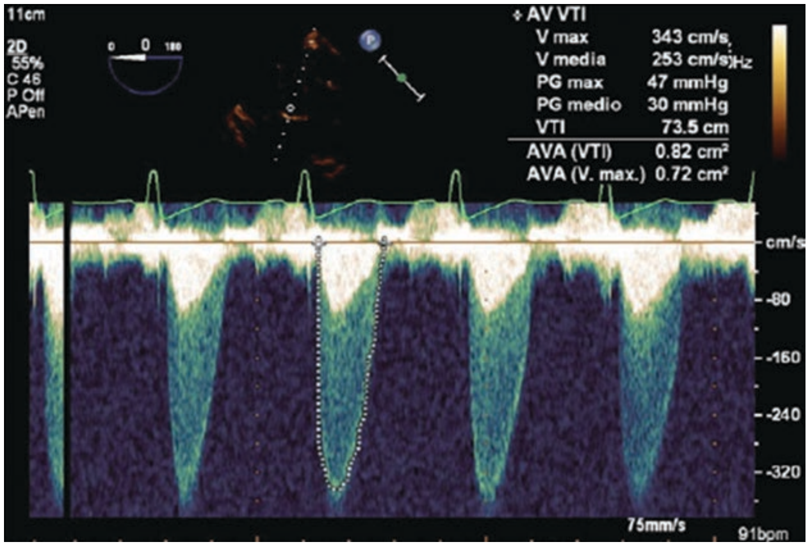

Fig. 1.13 Continuous-wave Doppler imaging

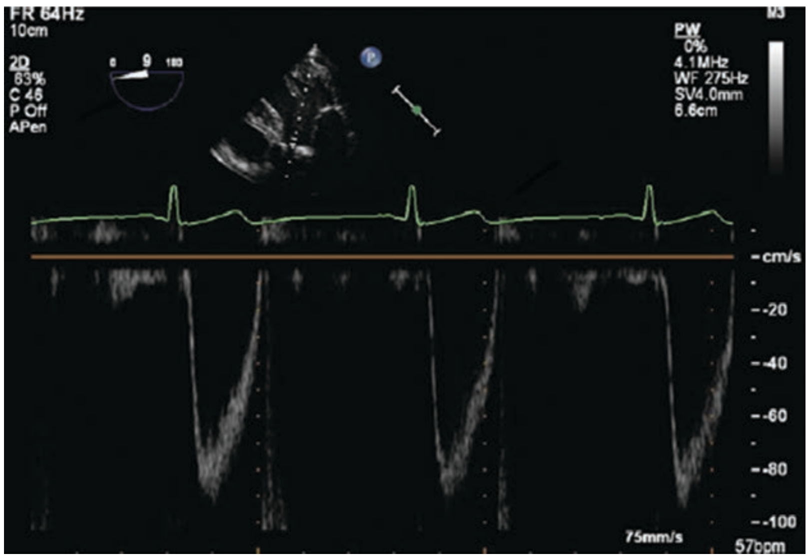

Fig. 1.14 Pulsed-wave Doppler echocardiography

The transducer must wait for the returning echo before sending out another ultrasound wave. The pulsed-wave Doppler technique samples velocities at a specific point (sample volume) of the ultrasound beam.

The number of pulses transmitted from a Doppler transducer each second is called the pulse repetition frequency (PRF). The sampling rate determinates the acquisition of information. If the Doppler shift frequency is higher than the PRF, the Doppler signal is displayed on the other side of the baseline. This is the alias artifact.

Aliasing occurs when the measured velocity is greater than half of the PRF (Figs. 1.15 and 1.16). This velocity is called the *Nyquist limit*.

There are some ways to improve the velocity performance of the pulsed-wave Doppler technique:

1. Decrease the distance between the transducer and the sample volume. Reducing the distance the ultrasound beam has to travel will increase the frequency of emission of the pulsed wave (PRF).

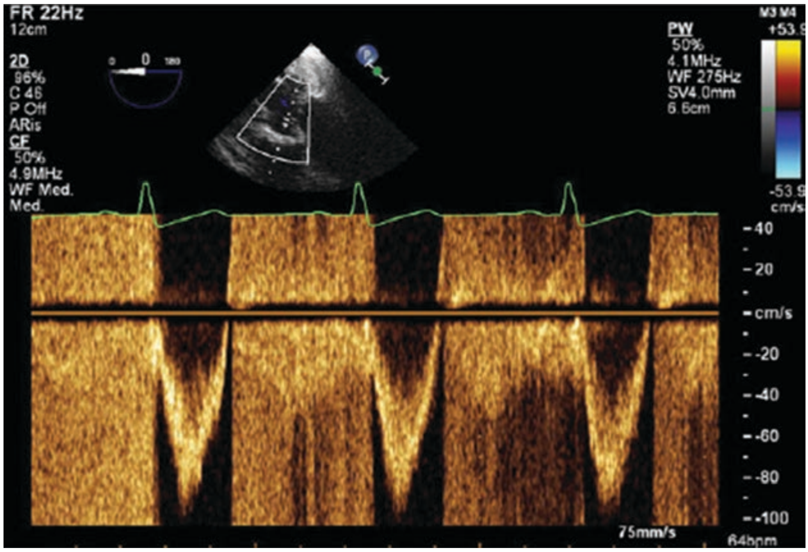

Fig. 1.15 Aliasing

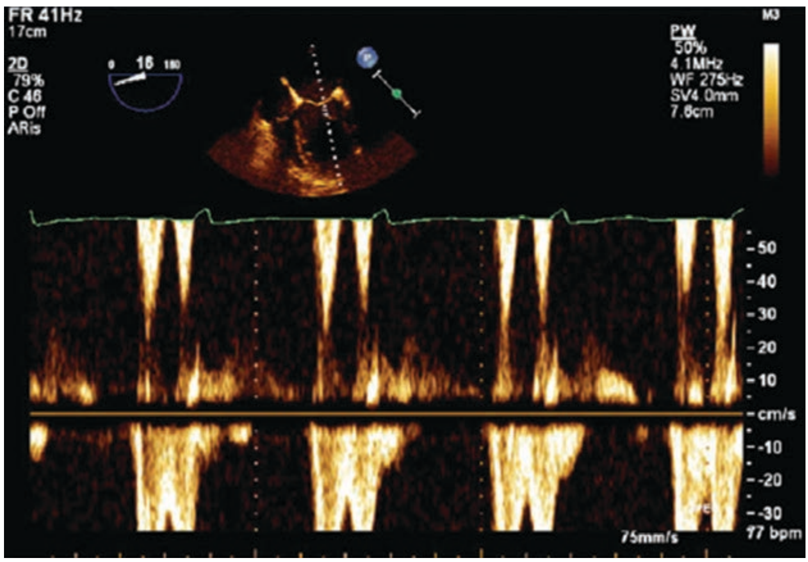

Fig. 1.16 Aliasing

2. Choose a low frequency of emission.
3. Set the baseline to display a greater range of velocities (Fig. 1.17).

In *tissue Doppler* imaging (Fig. 1.18), a low-pass filter is used to measure only the velocity of myocardial tissues. Tissue Doppler imaging uses a small pulsed-wave sample volume showing low velocity–high amplitude signals.

Color Doppler imaging combines a 2D image with a Doppler method to visualize the velocity of blood flow within an image plane. The Doppler shift of thousands of sample volumes displays the directions of the blood cells: blue for away from and red for toward the transducer (Fig. 1.19).

High-flow velocities are displayed in yellow (toward the transducer) and cyan (away from the transducer); green is used to visualize areas of turbulence. As with the pulsed-wave Doppler technique, the color flow Doppler technique suffers from the Nyquist limit, and aliasing can occur (Fig. 1.20).

Color M-mode Doppler imaging combines the spatiotemporal graphic representation of M-mode

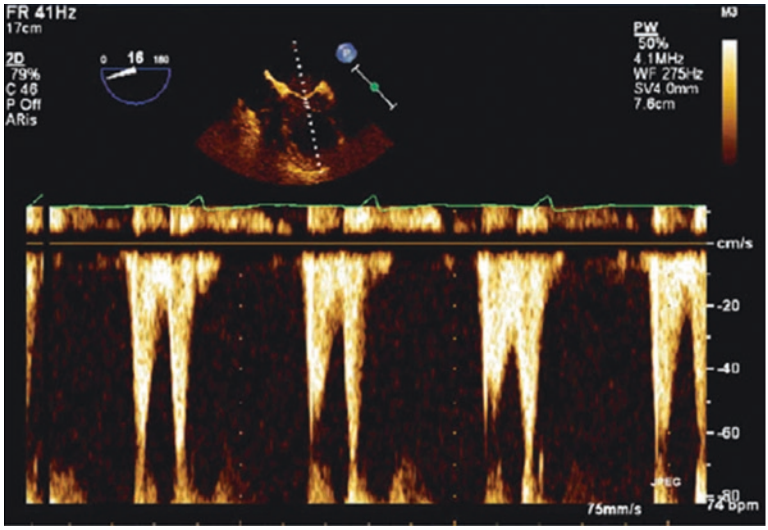

Fig. 1.17 Aliasing resolved by setting the right baseline

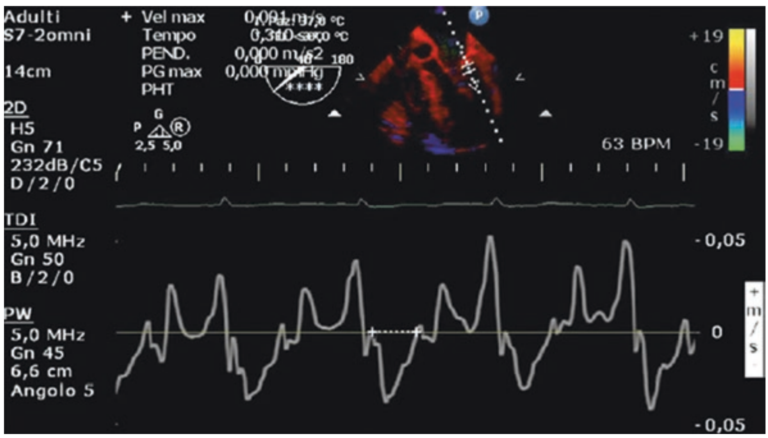

Fig. 1.18 Tissue Doppler imaging

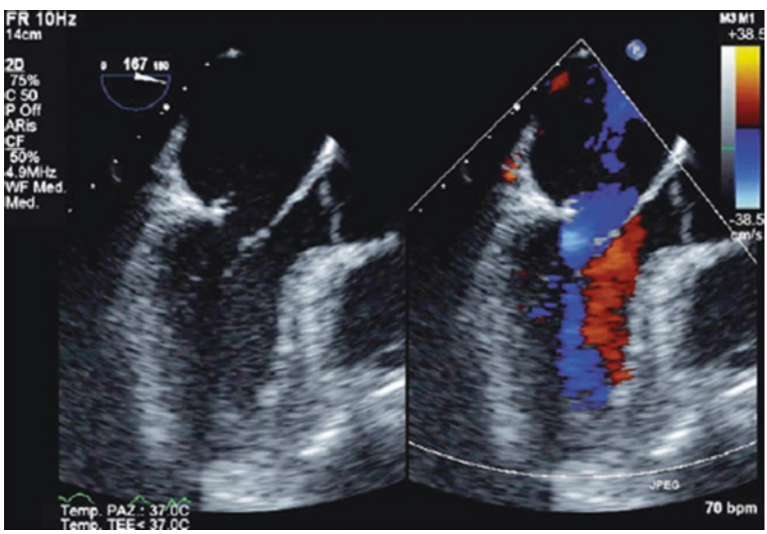

Fig. 1.19 Color Doppler imaging. *Blue* away from and *red* toward the transducer

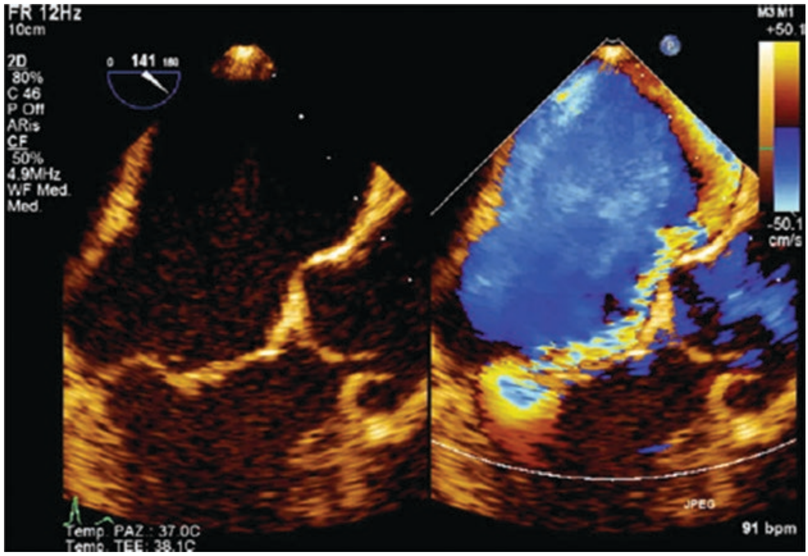

Fig. 1.20 Color Doppler aliasing

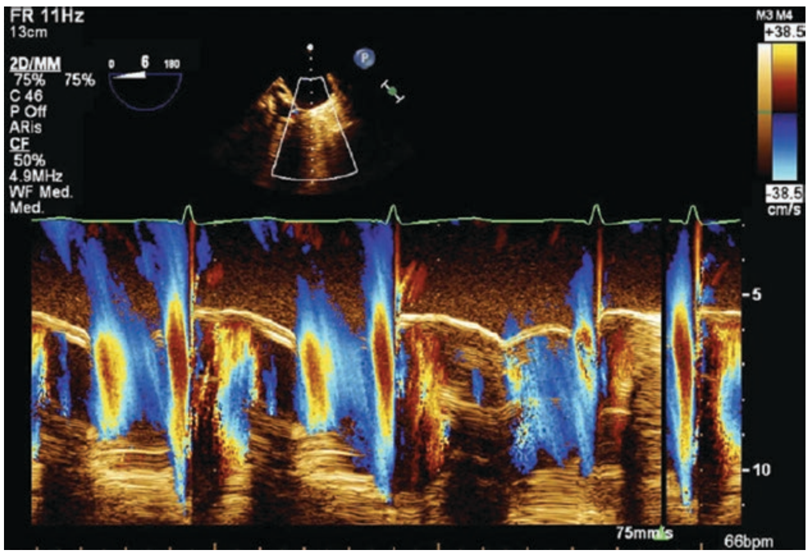

Fig. 1.21 Color M-mode imaging

and color codification. It shows at the same time a one-dimensional view of anatomic structures and color flow visualization. It is useful to assess transmitral flow (Fig. 1.21).

1.5 Use of the Ultrasound Machine: Optimizing the Picture

The image quality depends on the operator's skill and also on the adjustment of the ultrasound machine according to the features of the particular patient to be examined. The positioning of the patient and the probe is discussed in Chap. 2, together with all transthoracic views. First, it is essential to study well the instruction manual of the device at one's disposal to use it optimally.

1.5.1 Environment

The brightness of the environment where the examination is done should be reduced. The examination is performed in the ICU at the bedside, so it is preferable to have beds that can be easily arranged with Trendelenburg positioning, anti-Trendelenburg positioning, head and trunk lifting, and side tilting.

1.5.2 Ultrasonograph Setting

1.5.2.1 Electrocardiogram

Despite often being omitted to save time, it is important to always connect the electrocardiogram (ECG) wire of the ultrasound device to the patient electrodes. The ECG trace, recorded at the base of the display with a "marker" of time coinciding with the moving image, allows establishment of the phases of the cardiac cycle based on electrical activity of the heart, apart from monitoring the ECG. The mechanical systole usually begins immediately after the R wave and ends at about half of the T wave. The end of diastole coincides with the R-wave peak of the ECG (Fig. 1.22).

1.5.2.2 Probes

The probes used for adult echocardiography emit ultrasound with a frequency of about 3 MHz, whereas the probes used in pediatric echocardiography emit higher frequencies, from 5 to 7.5 MHz. These emissions represent the best compromise for use according to different types of patients, given that the higher the frequency, the better the image definition, but the lower the penetration of ultrasound into the tissues. Modern equipment

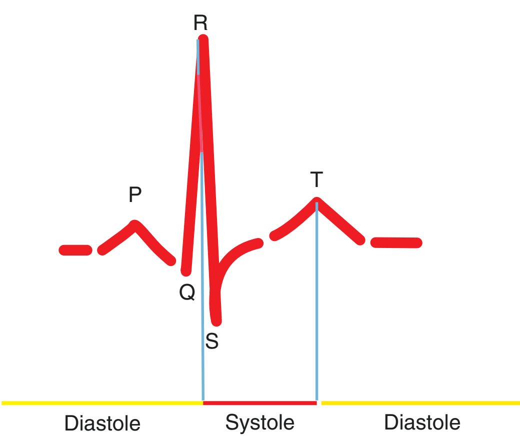

Fig. 1.22 ECG systolic and diastolic phases

Fig. 1.23 Ultrasound probes. From *left* to *right*: vascular and soft tissue linear probe, cardiac phased-array probe, abdominal convex probe

can produce sharper images through "tissue harmonic imaging": in a nutshell, the harmonics produced by the interaction of the ultrasound beam with the tissues are enhanced, and the fundamental harmonic frequencies are suppressed, resulting in better far-field quality. The image quality is better, but the very echo-reflective structures, such as pericardium and valves, can thus appear thicker than they really are. The probes have a touch and often light marker, defining the scanning plane and laterality. Figure 1.23 shows the various probes used for the study of the heart with transthoracic approaches, the vessels, and the thoracic and abdominal organs.

1.5.2.3 Sector Depth

The depth can be adjusted by the operator. The machine starts with a default standard depth so that the whole heart is displayed, but the depth of the field can be varied in order to position the structures of interest in the middle of the image. If the outer edges of the heart in the default image exceed the limits of the display, the heart as a whole or some part of it is certainly enlarged.

1.5.2.4 Width of the Scanning Beam

The maximum amplitude ensures that the most lateral structures are seen, but a reduction of the amplitude may sometimes be preferable to produce greater definition of the central structures; this is because a shorter time is needed to scan a narrower angle.

1.5.2.5 Gain

The construction of the image as grayscale or monochrome images depends on the intensity of the return signal, which depends on the distance traveled and on the reflective properties of the tissues encountered (see earlier). Therefore, the gain can be adjusted for different depths in order to compensate for the reduction of the return signal. This adjustment (time gain compensation), which is automatic in modern equipment, usually occurs through a system of levers that correspond to vertical depths of the field. Observing the display as the default, the operator improves the image manually by moving the levers that correspond to different levels of depth. In some devices there is also a system of horizontal adjustment for adjusting the image in the lateral fields (lateral gain compensation). However, these adjustments must be done with care since excess gain produces brighter images, leading to poor definition between close structures and even artifacts. In contrast, too dark images are produced with not enough gain, hiding some low-echo-reflective structures. Even though it depends somewhat on operator preference, a well-adjusted image (Fig. 1.24) is one that has:

- Fairly uniform intensity of solid structures
- A slight speckling in the dark cavities full of blood

1.5.2.6 Focus

The focus of the image is usually by default in the central part of the display, but one can move the focus to higher or lower levels for further research on particular structures.

1.5.2.7 Regulating Continuous-Wave Doppler and Pulsed-Wave Doppler Imaging

As already mentioned, continuous-wave Doppler imaging is used for the measurement of high flow rate in line with the cursor all along the stream to be examined. Pulsed-wave Doppler imaging is not suitable for high-speed flows but reproduces the flow in a specific area to be examined. The operator can adjust the gain of the Doppler signal. The optimal image is one that shows well the shape of the wave flow (changes in speed over time). By convention, the blood flow movement toward the transducer is represented above the baseline. In contrast, the movement away from the transducer is represented below the baseline. The scale of reproduction of the Doppler signal (y-axis) can be adjusted to avoid cutting high-speed-flow waves. The speed, usually 50 mm/s (x-axis), can be adjusted to better fit the times for special measures. The alignment of the ultrasound beam with the flow remains paramount for both continuous-wave and pulsed-wave interrogation. An angle of more than 30° between the blood flow

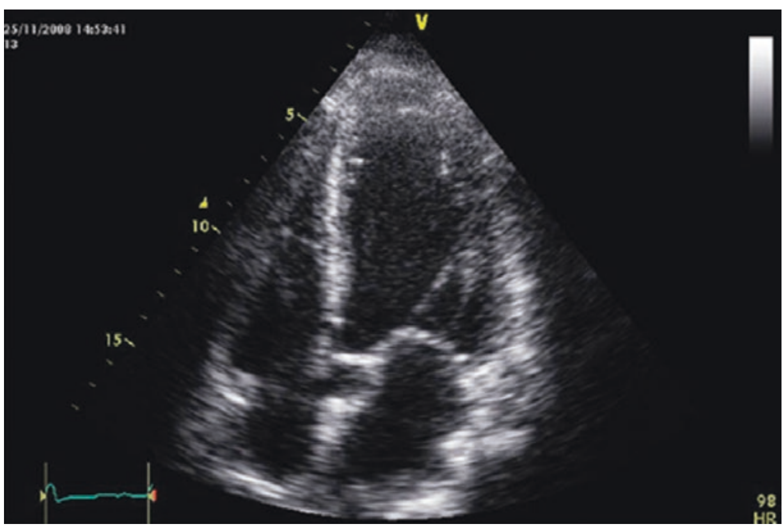

Fig. 1.24 Echocardiographic image (apical five-chamber view) with good contrast adjustment