Contents

1 Principles of Echocardiographic Image Acquisition and Doppler Analysis

BASIC PRINCIPLES

- Knowledge of basic ultrasound principles is needed for interpretation of images and Doppler data.
- Appropriate adjustment of instrument parameters is needed to obtain diagnostic information.

❖ KEY POINTS

- The appropriate ultrasound modality (two-dimensional [2D] or three-dimensional [3D] imaging, pulsed Doppler, color Doppler, etc.) is chosen for each type of needed clinical information.
- Current instrumentation allows modification of many parameters during data acquisition, such as depth, gain, harmonic imaging, wall filters, and so on.
- Artifacts must be distinguished from anatomic findings on ultrasound images.
- Accurate Doppler measurements depend on details of both blood flow interrogation and instrument acquisition parameters.

ULTRASOUND WAVES

- Ultrasound waves (Table 1.1) are mechanical vibrations with basic descriptors including:
  - Frequency (cycles per second = Hz, 1000 cycles/second = MHz)
  - Propagation velocity (about 1540 m/s in blood)
  - Wavelength (equal to the propagation velocity divided by frequency)
  - Amplitude (decibels [dBs])
- Ultrasound waves interact with tissues (Table 1.2) in four different ways:
  - Reflection (used to create ultrasound images)
  - Scattering (the basis of Doppler ultrasound)
  - Refraction (used to focus the ultrasound beam)
  - Attenuation (loss of signal strength in the tissue)

❖ KEY POINTS

- Tissue penetration is greatest with a lower frequency transducer (e.g., 2 to 3 MHz)
- Image resolution is greatest (about 1 mm) with a higher frequency transducer (e.g., 5 to 7.5 MHz) (Fig. 1.1)
- Amplitude ("loudness") is described using the logarithmic dB scale; a 6 dB change represents a doubling or halving of signal amplitude.
- Acoustic impedance depends on tissue density and the propagation velocity of ultrasound in that tissue.
- Ultrasound reflection occurs at smooth tissue boundaries with different acoustic impedances (such as between blood and myocardium). Reflection is greatest when the ultrasound beam is *perpendicular* to the tissue interface.
- Ultrasound scattering that occurs with small structures (such as red blood cells) is used to generate Doppler signals. Doppler velocity recordings are most accurate when the ultrasound beam is *parallel* to the blood flow direction.
- Refraction of ultrasound can result in imaging artifacts due to deflection of the ultrasound beam from a straight path.

TRANSDUCERS

- Ultrasound transducers use a piezoelectric crystal to alternately transmit and receive ultrasound signals (Fig. 1.2).
- Transducers are configured for specific imaging approaches—transthoracic, transesophageal, intracardiac, and intravascular (Table 1.3).

TABLE 1.1 Ultrasound Waves

	Definition	Examples	Clinical Implications
Frequency (f)	The number of cycles per second in an ultrasound wave: f = cycles/s = Hz	Transducer frequencies are measured in MHz (1,000,000 cycles/s). Doppler signal frequencies are measured in KHz (1000 cycles/s).	Different transducer frequencies are used for specific clinical applications, because the transmitted frequency affects ultrasound tissue penetration, image resolution, and the Doppler signal.
Velocity of propagation (c)	The speed that ultrasound travels through tissue	The average velocity of ultrasound in soft tissue about 1540 m/s.	The velocity of propagation is similar in different soft tissues (blood, myocardium, liver, fat, etc.) but is much lower in lung and much higher in bone.
Wavelength (λ)	The distance between ultrasound waves: $\lambda = c/f = 1.54/f$ (MHz)	Wavelength is shorter with a higher frequency transducer and longer with a lower frequency transducer.	Image resolution is greatest (about 1 mm) with a shorter wavelength (higher frequency). Depth of tissue penetration is greatest with a longer wavelength (lower frequency).
Amplitude (dB)	Height of the ultrasound wave or "loudness" measured in decibels (dB)	A log scale is used for dB. On the dB scale, 80 dB represents a 10,000-fold and 40 dB indicates a 100-fold increase in amplitude.	A very wide range of amplitudes can be displayed using a gray scale display for both imaging and spectral Doppler.

TABLE 1.2 Ultrasound Tissue Interaction

	Definition	Examples	Clinical Implications
Acoustic impedance (Z)	A characteristic of each tissue defined by tissue density (ϱ) and propagation of velocity (c) as: $z = \varrho \times c$	Lung has a low density and slow propagation velocity, whereas bone has a high density and fast propagation velocity. Soft tissues have smaller differences in tissue density and acoustic impedance.	Ultrasound is reflected from boundaries between tissues with differences in acoustic impedance (e.g., blood versus myocardium).
Reflection	Return of ultrasound signal to the transducer from a smooth tissue boundary	Reflection is used to generate 2D cardiac images.	Reflection is greatest when the ultrasound beam is perpendicular to the tissue interface.
Scattering	Radiation of ultrasound in multiple directions from a small structure, such as blood cells	The change in frequency of signals scattered from moving blood cells is the basis of Doppler ultrasound.	The amplitude of scattered signals is 100 to 1000 times less than reflected signals.
Refraction	Deflection of ultrasound waves from a straight path due to differences in acoustic impedance	Refraction is used in transducer design to focus the ultrasound beam.	Refraction in tissues results in double image artifacts.
Attenuation	Loss in signal strength due to absorption of ultrasound energy by tissues	Attenuation is frequency dependent with greater attenuation (less penetration) at higher frequencies.	A lower frequency transducer may be needed for apical views or in larger patients on transthoracic imaging.
Resolution	The smallest resolvable distance between two specular reflectors on an ultrasound image	Resolution has three dimensions—along the length of the beam (axial), lateral across the image (azimuthal), and in the elevational plane.	Axial resolution is most precise (as small as 1 mm), so imaging measurements are best made along the length of the ultrasound beam.

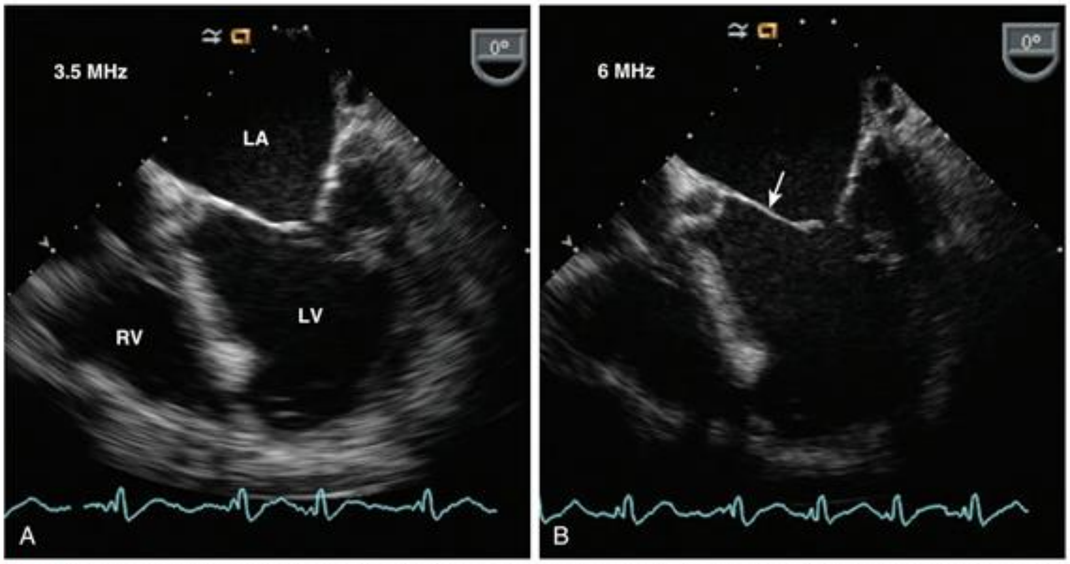

Fig 1.1 The effect of transducer frequency on penetration and resolution. In this transesophageal 4-chamber view recorded at a transmitted frequency of 3.5 MHz (A) and 6 MHz (B), the higher frequency transducer provides better resolution—for example, the mitral leaflets *(arrow)* look thin, but the depth of penetration of the signal is very poor, so the apical half of the LV is not seen. With the lower frequency transducer, improved tissue penetration provides a better image of the LV apex but image resolution is poorer, with the mitral leaflets looking thicker and less well defined.

- The basic characteristics of a transducer are:
  - Transmission frequency (from 2.5 MHz for transthoracic to 20 MHz for intravascular ultrasound)
  - Bandwidth (range of frequencies in the transmitted ultrasound pulse)
  - Pulse repetition frequency (the number of transmission-receive cycles per second)
  - Focal depth (depends on beam shape and focusing)
  - Aperture (size of the transducer face or "footprint")
  - Power output

❖ KEY POINTS

- The time delay between transmission of an ultrasound burst and detection of the reflected wave indicates the depth of the tissue reflector.
- The pulse repetition frequency is an important factor in image resolution and frame rate.
- A shorter transmitted pulse length results in improved depth (or axial) resolution.
- A wider bandwidth provides better resolution of structures distant from the transducer.
- The shape of the ultrasound beam depends on several complex factors. Each type of transducer focuses the beam at a depth appropriate for the clinical application. Some transducers allow adjustment of focal depth.

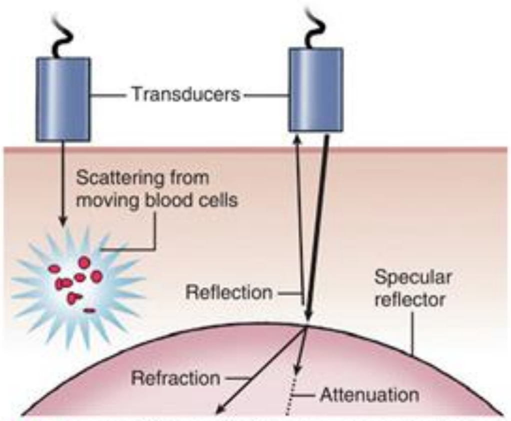

Fig 1.2 Diagram of the interaction between ultrasound and body tissues. Doppler analysis is based on the scattering of ultrasound in all directions from moving blood cells with a resulting change in frequency of the ultrasound received at the transducer. 2D imaging is based on reflection of ultrasound from tissue interfaces (specular reflectors). Attenuation limits the depth of ultrasound penetration. Refraction, a change in direction of the ultrasound wave, results in imaging artifacts. *(From Otto, CM:* Textbook of clinical echocardiography, *ed 6, Elsevier, 2018, Philadelphia.)*

- A smaller aperture is associated with a wider beam width; however, the smaller "footprint" may allow improved angulation of the beam in the intercostal spaces. This is most evident clinically with a dedicated non-imaging continuous-wave (CW) Doppler transducer.

ULTRASOUND IMAGING

Principles

- The basic ultrasound imaging modalities are:
    - M-mode—a graph of depth versus time
    - 2D—a sector scan in a tomographic image plane with real-time motion
    - 3D—a selected cutaway real-time image in a 3D display format or a 3D volume of data (see Chapter 4)
- System controls for 2D imaging typically include:
    - Power output (transmitted ultrasound energy)
    - Gain (amplitude of the received signal)
    - Time gain compensation (differential gain along the ultrasound beam)
    - Depth of the image (affects pulse repetition frequency and frame rate)
    - Gray scale/dynamic range (degree of contrast in the images)

❖ KEY POINTS

- M-mode recordings allow identification of very rapid intracardiac motion, because the sampling rate is about 1800 times per second compared to a 2D frame rate of 30 frames per second (Fig. 1.3).

TABLE 1.3 Ultrasound Transducers

	Definition	Examples	Clinical Implications
Type	Transducer characteristics and configuration Most cardiac transducers use phased array of piezoelectric crystals	Transthoracic (adult and pediatric) Non-imaging CW Doppler 3D echocardiography TEE Intracardiac	Each transducer type is optimized for a specific clinical application. More than one transducer may be needed for a full examination.
Transmission frequency	The central frequency emitted by the transducer	Transducer frequencies vary from 2.5 MHz for transthoracic echo to 20 MHz for intravascular imaging.	A higher frequency transducer provides improved resolution but less penetration. Doppler signals are optimal at a lower transducer frequency than used for imaging.
Power output	The amount of ultrasound energy emitted by the transducer	An increase in transmitted power increases the amplitude of the reflected ultrasound signals.	Excessive power output may result in bioeffects measured by the mechanical and thermal indexes.
Bandwidth	The range of frequencies in the ultrasound pulse	Bandwidth is determined by transducer design.	A wider bandwidth allows improved axial resolution for structures distant from the transducer.
Pulse (or burst) length	The length of the transmitted ultrasound signal	A higher frequency signal can be transmitted in a shorted pulse length compared with a lower frequency signal.	A shorter pulse length improves axial resolution.
Pulse repetition frequency (PRF)	The number of transmission-receive cycles per second	The PRF decreases as imaging (or Doppler) depth increases because of the time needed for the signal to travel from and to the transducer.	Pulse repetition frequency affects image resolution and frame rate (particularly with color Doppler)
Focal depth	Beam shape and focusing are used to optimize ultrasound resolution at a specific distance from the transducer	Structures close to the transducer are best visualized with a short focal depth, distant structures with a long focal depth.	The length and site of a transducer's focal zone is primarily determined by transducer design, but adjustment during the exam may be possible.
Aperture	The surface of the transducer face where ultrasound is transmitted and received	A small non-imaging CW Doppler transducer allows optimal positioning and angulation of the ultrasound beam.	A larger aperture allows a more focused beam. A smaller aperture allows improved transducer angulation on TTE imaging.

- Ultrasound imaging resolution is more precise along the length of the ultrasound beam (axial resolution) compared with lateral (side to side) or elevational ("thickness" of the image plane) resolution.
- Lateral resolution decreases with increasing distance from the transducer (Fig. 1.4).
- Harmonic imaging improves endocardial definition and reduces near-field and side-lobe artifacts (Fig. 1.5).

Imaging Artifacts

- Common imaging artifacts result from:
  - A low signal-to-noise ratio
  - Acoustic shadowing
  - Reverberations
  - Beam width
  - Lateral resolution
  - Refraction
  - Range ambiguity
  - Processing

❖ KEY POINTS

- A shadow occurs distal to a strong ultrasound reflector because the ultrasound wave does not penetrate past the reflector (Fig. 1.6).
- Signals originating from the edges of the ultrasound beam or from side lobes can result in imaging or Doppler artifacts.
- Deviation of the ultrasound beam from a straight pathway due to refraction in the tissue results in the structure appearing in the incorrect location across the sector scan (Fig. 1.7).
- Ultrasound reflected back and forth between two strong reflectors creates a reverberation artifact.
- Reflected ultrasound signals received at the transducer are assumed to originate from the preceding transmitted pulse. Signals from very deep structures or signals that have been re-reflected will be displayed at one-half or twice the actual depth of origin.

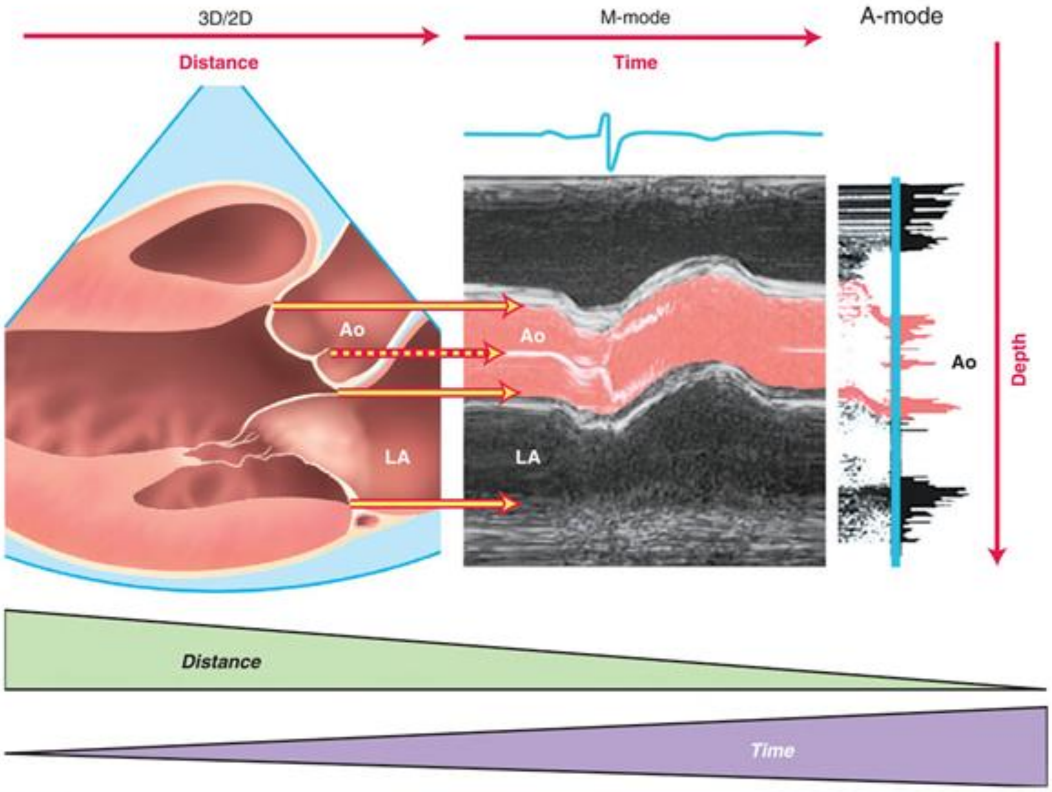

Fig 1.3 **3D, 2D, M-mode, and A-mode recordings of aortic valve motion.** This illustration shows the relationship between the 3D and 2D long-axis image of the aortic valve *(left)*, which shows distance in both the vertical and horizontal directions, M-mode recording of aortic root *(Ao)*, *LA*, and aortic valve motion, which shows depth versus time *(middle)* and A-mode recording *(right)*, which shows depth only (with motion seen on the video screen). Spatial relationships are best shown with 3D/2D, but temporal resolution is higher with M-mode and A-mode imaging. *(From Otto, CM: Textbook of clinical echocardiography, ed 6, Elsevier, 2018, Philadelphia.)*

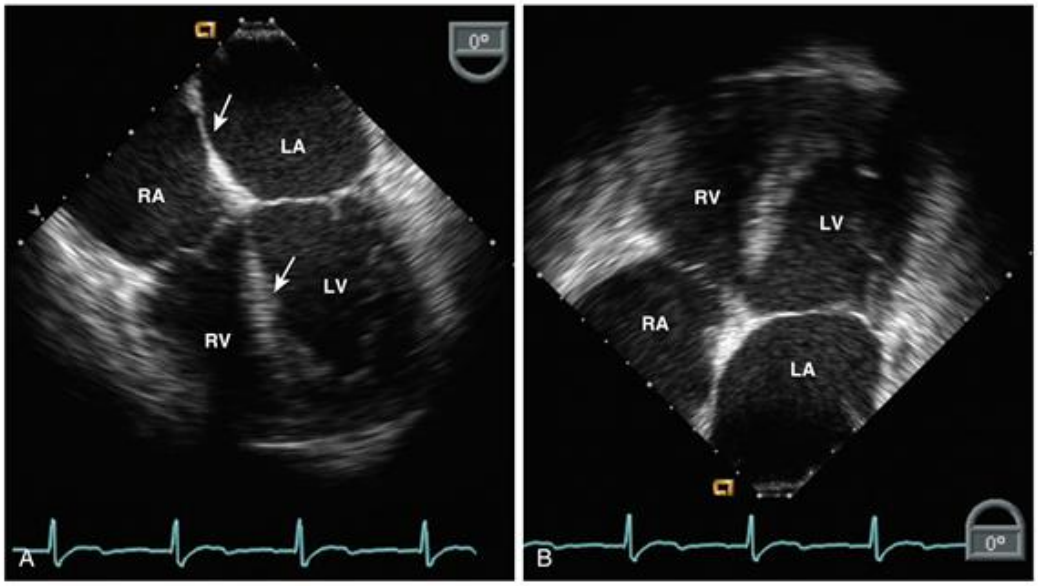

Fig 1.4 **Lateral resolution with ultrasound decreases with the distance of the reflector from the transducer.** (A) In this TEE image oriented with the origin of the ultrasound signal at the top of the image, thin structures close to the transducer, such as the atrial septum *(upper arrow)*, appear as a dot because lateral resolution is optimal at this depth. Reflections from more distant structures, such as the ventricular septum *(lower arrow)*, appear as a broad line due to poor lateral resolution. (B) When the image is oriented with the transducer at the bottom of the image, the effects of depth on lateral resolution are more visually apparent. The standard orientation for echocardiography with the transducer as the top of the image is based on considerations of ultrasound physics, not on cardiac anatomy.

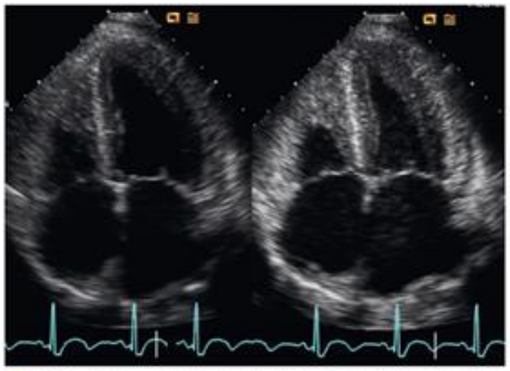

Fig 1.5 **Harmonic imaging compared with fundamental frequency imaging.** Harmonic imaging improves identification of the LV endocardial border, as seen in this apical 4-chamber view recorded with a 4-MHz transducer using *(left)* fundamental frequency imaging and *(right)* harmonic imaging.

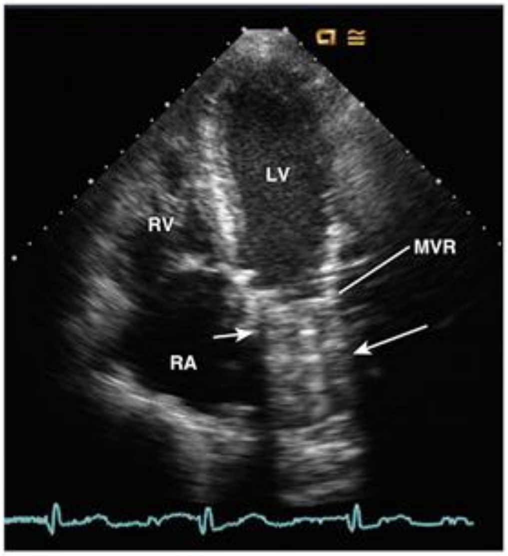

Fig 1.6 **Acoustic shadowing and reverberations.** This apical 4-chamber view in a patient with a mechanical mitral valve replacement *(MVR)* illustrates the shadowing (dark area, *small arrow*) and reverberations (white band of echoes, *large arrow*) that obscure structures (in this case, the left atrium) distal to the valve.