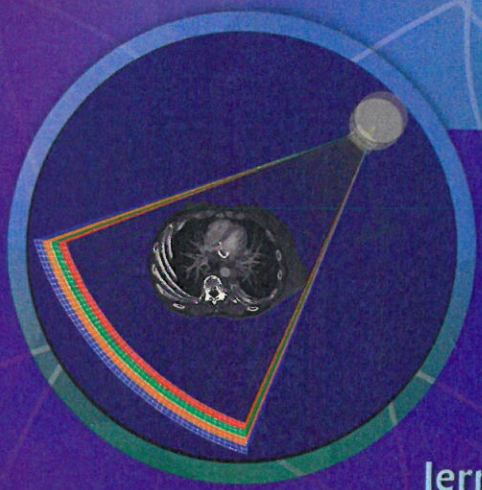
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The Essential Physics of Medical Imaging

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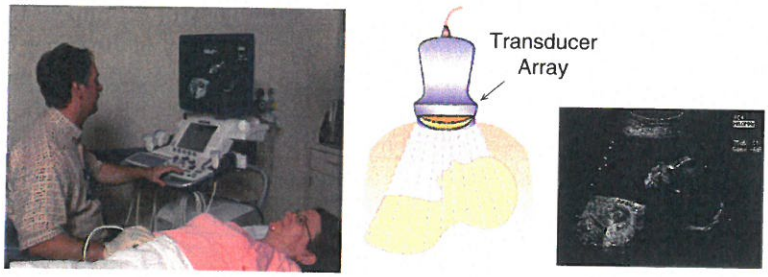
CHAPTER 14

Ultrasound

Ultrasound is the term that describes sound waves of frequencies exceeding the range of human hearing and their propagation in a medium. Medical diagnostic ultrasound is a modality that uses ultrasound energy and the acoustic properties of the body to produce an image from stationary and moving tissues. In ultrasound imaging, a short pulse of mechanical energy is delivered to the tissues. The pulse travels at the speed of sound, and with changes in the tissue acoustic properties, a fraction of the pulse is reflected as an echo that returns to the source. Collection of the echoes over time and recording of the echo amplitudes provide information about the tissues along the path of travel. Repeating the process hundreds of times with a small incremental change in the direction of the pulse interrogates a volume, from which a gray-scale tomographic image can be synthesized. Generation of the sound pulses and detection of the echoes are accomplished with a transducer, which also directs the ultrasound pulse along a linear path through the patient. Along a given beam path, the depth of an echo-producing structure is determined from the time between the pulse emission and the echo return, and the amplitude of the echo is encoded as a gray-scale value (Fig. 14-1). In addition to two-dimensional (2D) tomographic imaging, ultrasound provides anatomical distance and volume measurements, motion studies, blood velocity measurements, and 3D imaging.

Historically, medical uses of ultrasound came about shortly after the close of World War II, derived from underwater sonar research. Initial clinical applications monitored changes in the propagation of pulses through the brain to detect intracerebral hematoma and brain tumors based upon the displacement of the midline. Ultrasound rapidly progressed through the 1960s from simple "A-mode" scans to "B-mode" applications and compound "B-scan" images using analog electronics. Advances in equipment design, data acquisition techniques, and data processing capabilities have led to electronic transducer arrays, digital electronics, and real-time image display. This progress is changing the scope of ultrasound and its applications in diagnostic radiology and other areas of medicine. High-resolution, real-time imaging, harmonic imaging, 3D data acquisition, and power Doppler are a few of the innovations introduced into clinical practice. Contrast agents for better delineation of the anatomy, measurement of tissue perfusion, precise drug delivery mechanisms, and determination of elastic properties of the tissues are examples of current research.

The characteristics, properties, and production of ultrasound are described, followed by modes of ultrasound interaction, instrumentation, and image acquisition methods. Signal processing and image display methods show how the ultrasound image is produced, and how evaluation of the blood velocity through Doppler techniques is achieved. Types of image artifacts that can occur, quality control (QC) tests, and potential ultrasound bioeffects complete the overview.

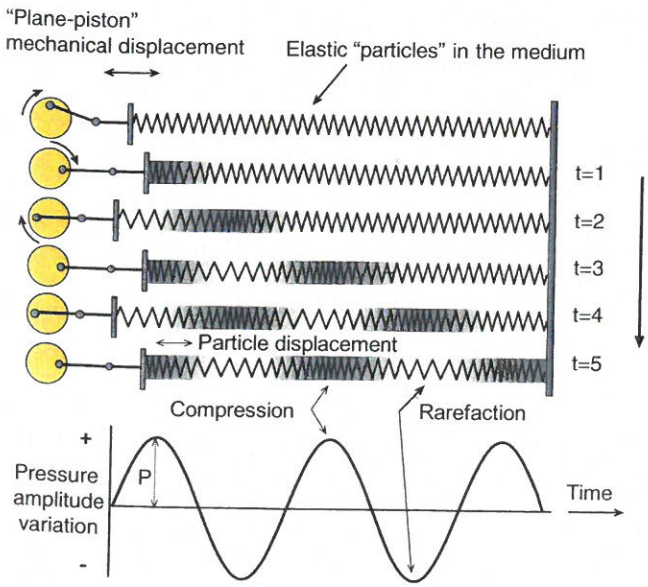


■ **FIGURE 14-1** A major use of ultrasound is the acquisition and display of the acoustic properties of tissues. A transducer array (transmitter and receiver of ultrasound pulses) directs sound waves into the patient, receives the returning echoes, and converts the echo amplitudes into a 2D tomographic image using the ultrasound acquisition system. Exams requiring a high level of safety such as obstetrics are increasing the use of ultrasound in diagnostic radiology. Photo credit: Emi Manning UC Davis Health System.

14.1 Characteristics of Sound

Propagation of Sound

Sound is mechanical energy that propagates through a continuous, elastic medium by the compression and rarefaction of "particles" that comprise it. A simple model of an elastic medium is that of a spring, as shown in Figure 14-2. Compression is caused by a mechanical deformation induced by an external force (such as a plane piston), with a resultant increase in the pressure of the medium. Rarefaction occurs following the compression event—as the backward motion of the piston reverses the force, the compressed particles transfer their energy to adjacent particles with a subsequent reduction in the local pressure amplitude. The section of the spring in contact with the piston then stretches and becomes decompressed (a rarefaction), while the compression continues to travel forward. With continued back-and-forth motion of the piston, a



■ **FIGURE 14-2** Ultrasound energy is generated by a mechanical displacement in compressible medium, which is modeled as an elastic spring. Energy propagation is shown as a function of time, resulting in areas of compression and rarefaction with corresponding variations in positive and negative pressure amplitude, *P* (lower diagram). Particles in the medium have only minor displacement as the wave front passes.

series of compressions and rarefactions propagate through the continuous medium (the spring). Any change with respect to position in the elasticity of the spring disturbs the wave train and causes some or all of the energy to be reflected. While the medium itself is necessary for mechanical energy transfer (i.e., sound propagation), the constituent "particles" of the medium act only to transfer mechanical energy; these particles experience only very small back-and-forth displacements. Energy propagation occurs as a wave front in the direction of energy travel, known as a longitudinal wave.

Wavelength, Frequency, and Speed

The *wavelength* (λ) of the ultrasound energy is the distance (usually expressed in units of *mm* or μm) between compressions or rarefactions, or between any two points that repeat on the sinusoidal wave of pressure amplitude (see Fig. 14-2). The *frequency* (*f*) is the number of times the wave oscillates through one cycle each second (*s*). Sound waves with frequencies less than 15 cycles per second (Hz) are called infrasound, and the range between 15 Hz and 20 kHz comprises the audible acoustic spectrum. Ultrasound represents the frequency range above 20 kHz. Medical ultrasound uses frequencies in the range of 2 to 10 MHz, with specialized ultrasound applications up to 50 MHz. The *period* is the time duration of one wave cycle and is equal to $1/f$, where f is expressed in cycles/s. The *speed of sound* is the distance traveled by the wave per unit time and is equal to the wavelength divided by the period. Since period and frequency are inversely related, the relationship between speed, wavelength, and frequency for sound waves is

$$c = \lambda f$$

where c (*m/s*) is the speed of sound in the medium, λ (*m*) is the wavelength, and f (cycles/s) is the frequency.

The speed of sound is dependent on the propagation medium and varies widely in different materials. The wave speed is determined by the ratio of the bulk modulus (B) (a measure of the stiffness of a medium and its resistance to being compressed) and the density (ρ) of the medium:

$$c = \sqrt{\frac{B}{\rho}}$$

SI units are $kg/(m\text{-}s^2)$, kg/m^3, and *m/s* for B, ρ and c, respectively. A highly compressible medium, such as air, has a low speed of sound, while a less compressible medium, such as bone, has a higher speed of sound. A less dense medium has a higher speed of sound than a denser medium (e.g., dry air versus humid air). The speeds of sound in materials encountered in medical ultrasound are listed in Table 14-1. Of major importance is the average speed for "soft tissue" (1,540 *m/s*), fatty tissue (1,450 *m/s*), and air (330 *m/s*). The difference in the speed of sound at tissue boundaries is a fundamental property that generates echoes (and contrast) in an ultrasound image. Medical ultrasound machines use a speed of sound of 1,540 *m/s* when determining localization of reflectors and creating the acoustic image. The speed of sound in soft tissue can be expressed in different units; besides 1,540 *m/s*, most common are 154,000 *cm/s* and 1.54 *mm/*μ*s*, which are often helpful in simplifying conversion factors or estimating values.

The ultrasound frequency is unaffected by changes in sound speed as the acoustic beam propagates through different media. Thus, the ultrasound wavelength is dependent on the medium. Examples of the wavelength change as a function of frequency and propagation medium are given below.

TABLE 14-1 DENSITY AND SPEED OF SOUND IN TISSUES AND MATERIALS FOR MEDICAL ULTRASOUND

MATERIAL	DENSITY (kg/m³)	c (m/s)	c (mm/μs)
Air	1.2	330	0.33
Lung	300	600	0.60
Fat	924	1,450	1.45
Water	1,000	1,480	1.48
"Soft Tissue"	1,050	1,540	1.54
Kidney	1,041	1,565	1.57
Blood	1,058	1,560	1.56
Liver	1,061	1,555	1.55
Muscle	1,068	1,600	1.60
Skull bone	1,912	4,080	4.08
PZT	7,500	4,000	4.00

EXAMPLE: A 2-MHz beam has a wavelength in soft tissue of

$$\lambda = \frac{c}{f} = \frac{1540\ m/s}{2\times10^6/s} = 770\times10^{-6}\ m = 7.7\times10^{-4}\ m\times1000\frac{mm}{m} = 0.77\ mm$$

A 10-MHz ultrasound beam has a corresponding wavelength in soft tissue of

$$= \frac{1540\ m/s}{10\times10^6/s} = 154\times10^{-6}\ m = 1.54\times10^{-4}\ m\times1000\frac{mm}{m} \cong 0.15\ mm$$

Higher frequency sound has shorter wavelength, as shown in Figure 14-3.

EXAMPLE: A 5-MHz beam travels from soft tissue into fat. Calculate the wavelength in each medium, and determine the percent wavelength change.

In soft tissue,

$$\lambda = \frac{c}{f} = \frac{1540\ m/s}{5\times10^6/s} = 3.08\times10^{-6}\ m \cong 0.31 mm$$

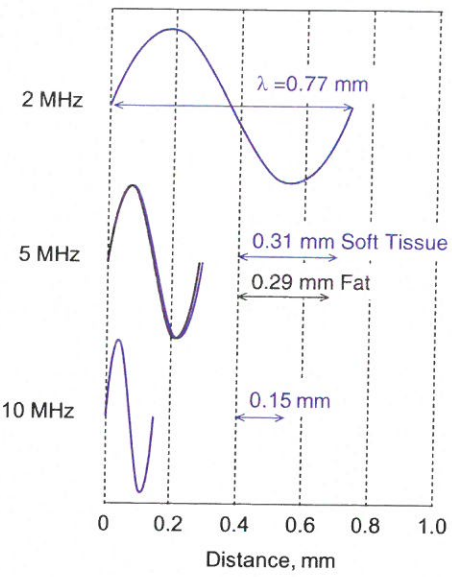
In fat,

$$= \frac{1450\ m/s}{5\times10^6/s} = 2.9\times10^{-6}\ m \cong 0.29\ mm$$

A *decrease* in wavelength of 5.8% occurs in going from soft tissue into fat, due to the differences in the speed of sound. This is depicted in Figure 14-3 for a 5-MHz ultrasound beam.

The wavelength in mm in soft tissue can be calculated directly by dividing the speed of sound in mm/μs (c = 1,540 m/s = 1.54 mm/μs) by the frequency in MHz, as

$$\lambda(mm,\ \text{soft tissue}) = \frac{c}{f} = \frac{1.54\ mm/\mu s}{f(MHz)} = \frac{1.54\ mm/10^{-6}\,s}{f(10^6/s)} = \frac{1.54\ mm}{f\ (\text{MHz})}$$



■ **FIGURE 14-3** Ultrasound wavelength is determined by the frequency and the speed of sound in the propagation medium. Wavelengths in soft tissue are calculated for 2-, 5-, and 10-MHz ultrasound sources for soft tissue (*blue*). A comparison of wavelength in fat (*purple*) to soft tissue at 5 MHz is also shown.

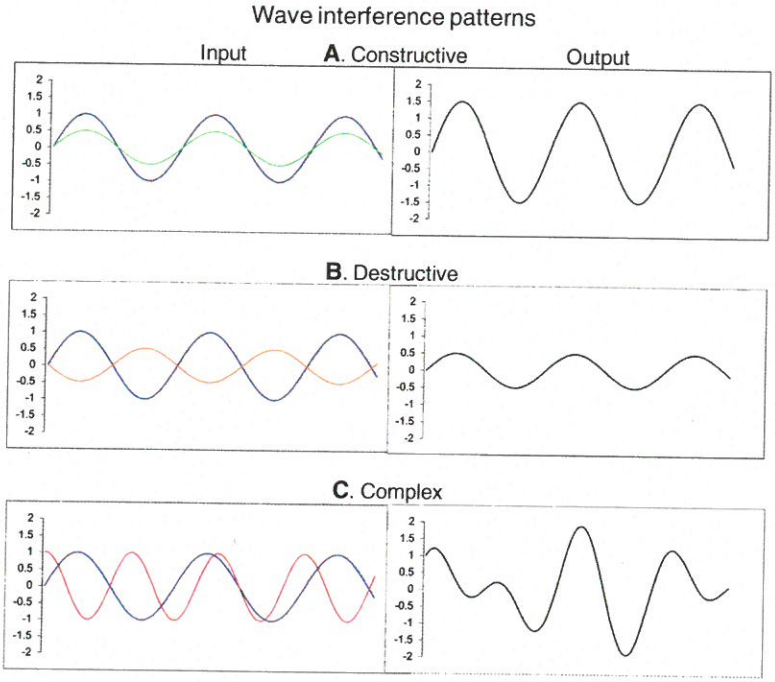
Thus, it can quickly be determined that the wavelength of a 10-MHz beam is 0.154 mm, and that of a 1-MHz beam is 1.54 mm by using this simple change in units.

The spatial resolution of the ultrasound image and the attenuation of the ultrasound beam energy depend on the wavelength and frequency, respectively. Ultrasound wavelength affects the spatial resolution achievable along the direction of the beam. A high-frequency ultrasound beam (small wavelength) provides better resolution and image detail than a low-frequency beam; however, the depth of beam penetration is significantly reduced at higher frequency. Ultrasound frequencies used for imaging are therefore selected according to the imaging application. For body parts requiring greater travel distance of the sound waves (e.g., abdominal imaging), lower frequency ultrasound is used (3.5 to 5 MHz) to image structures at significant depths. For small body parts or organs close to the skin surface (e.g., thyroid, breast), higher frequency ultrasound is selected (7.5 to 10 MHz). Most medical imaging applications use ultrasound frequencies in the range of 2 to 10 MHz.

Modern ultrasound equipment consists of multiple sound transmitters aligned in an array that create sound beams independently. Interaction of two or more separate ultrasound beams in a medium can result in *constructive* and/or *destructive wave interference*, as shown in Figure 14-4. The amount of constructive or destructive interference depends on several factors, but the most important are the phase (position of the periodic wave with respect to a reference point) and amplitude of the interacting beams. With phase differences occurring because of the distances between individual transmitters and time of transmitter excitation, all of the individual beams' interactions can generate complex interference patterns, which are very important in shaping and steering the composite ultrasound beam.

Pressure, Intensity, and the *dB* Scale

Sound energy causes particle displacements and variations in local pressure amplitude, P, in the propagation medium. Pressure amplitude is defined as the peak maximum or peak minimum value from the average pressure on the medium in the absence of a sound wave. In the case of a symmetrical waveform, the positive and



■ **FIGURE 14-4** Interaction of waves in a propagation medium, plotted as amplitude versus time. **A.** Constructive interference occurs with two ultrasound waves of the same frequency and phase, resulting in a higher amplitude output wave. **B.** Destructive interference occurs with the waves 180 degrees out-of-phase, resulting in a lower amplitude output wave. **C.** Complex interference occurs when waves of slightly different frequency interact, resulting in an output waveform of higher and lower amplitude.

negative pressure amplitudes are equal (lower portion of Fig. 14-2); however, in most diagnostic ultrasound applications, the compressional amplitude significantly exceeds the rarefactional amplitude. The SI unit of pressure is the Pascal (Pa), defined as one Newton per square meter (N/m^2). The average atmospheric pressure on earth at sea level is approximately equal to 100,000 Pa. Diagnostic ultrasound beams typically deliver peak pressure levels that exceed ten times the earth's atmospheric pressure, or about one MPa (mega Pascal).

Intensity, I, is the amount of power (energy per unit time) per unit area and is proportional to the square of the pressure amplitude, $I \propto P^2$. Medical diagnostic ultrasound intensity levels are described in units of milliwatts/cm^2—the amount of energy per unit time per unit area. The absolute intensity level depends upon the method of ultrasound production (e.g., pulsed or continuous—a discussion of these parameters can be found in Section 14.11 of this chapter). Relative intensity and pressure levels are described as a logarithmic ratio, the *decibel* (*dB*). The relative intensity in *dB* for intensity ratios is calculated as

$$\text{relative intensity (dB)} = 10 \log \left(\frac{I_2}{I_1} \right)$$

where I_1 and I_2 are intensity values that are compared as a *relative measure*, and "log" denotes the base 10 logarithm. In diagnostic ultrasound, the ratio of the intensity of the incident pulse to that of the returning echo can span a range of one million times or more. The logarithm function compresses the large and expands the small ratios into a more manageable number range. An intensity ratio of 10^6 (e.g., an incident intensity

TABLE 14-2 INTENSITY RATIO AND CORRESPONDING DECIBEL VALUES

INTENSITY RATIO		DECIBELS (*dB*)
I_2/I_1	LOG (I_2/I_1)	
1	0	0
2	0.3	3
10	1	10
100	2	20
10,000	4	40
1,000,000	6	60
0.5	−0.3	−3
0.01	−2	−20
0.0001	−4	−40
0.000001	−6	−60

one million times greater than the returning echo intensity) is equal to 60 *dB*, whereas an intensity ratio of 10^2 is equal to 20 *dB*. A change of 10 in the *dB* scale corresponds to an order of magnitude (10 times) change in intensity; a change of 20 corresponds to two orders of magnitude (100 times) change, and so forth. When the intensity ratio is greater than one (e.g., the incident ultrasound intensity to the detected echo intensity), the *dB* values are positive; when less than one, the *dB* values are negative. A loss of 3 *dB* (−3 *dB*) represents a 50% loss of signal intensity. The tissue thickness that reduces the ultrasound intensity by 3 *dB* is considered the "half-value" thickness (HVT). In Table 14-2, intensity ratios, logarithms, and the corresponding intensity dB values are listed.

EXAMPLE: Calculate the remaining intensity of a 100 mW ultrasound pulse that loses 30 dB while traveling through tissue.

$$\text{relative intensity } (dB) = 10 \log \frac{I_2}{I_1}; \quad \text{defining equation}$$

$$-30\ dB = 10 \log \frac{I_2}{100\ \text{mW}}; \quad \text{divide each side of the equation by 10:}$$

$$-3 = \log \frac{I_2}{100\ \text{mW}}; \quad \text{exponentiate using base 10}$$

$$10^{-3} = \frac{I_2}{100\ \text{mW}}; \quad \text{solve for } I_2:$$

$$I_2 = 0.001 \times 100\ \text{mW} = 0.1\ \text{mW}$$

14.2 Interactions of Ultrasound with Matter

Ultrasound interactions are determined by the acoustic properties of matter. As ultrasound energy propagates through a medium, interactions include reflection, refraction, scattering, and absorption. *Reflection* occurs at tissue boundaries where there is a difference in the *acoustic impedance* of adjacent materials. When the incident beam is perpendicular to the boundary, a fraction of the beam (an echo) returns directly

back to the source; the transmitted fraction of the beam continues in the initial direction. For non-normal incidence, the incident angle is that made relative to normal incidence; the reflected angle is equal to the incident angle. *Refraction* describes the change in direction of the transmitted ultrasound energy with nonperpendicular incidence. *Scattering* occurs by reflection or refraction, usually by small particles within the tissue medium, causes the beam to diffuse in many directions, and gives rise to the characteristic texture and gray scale in the acoustic image. *Attenuation* refers to the loss of intensity of the ultrasound beam from absorption and scattering in the medium. *Absorption* is the process whereby acoustic energy is converted to heat energy, whereby, sound energy is lost and cannot be recovered.

Acoustic Impedance

The acoustic impedance (Z) of a material is defined as

$$Z = \rho c$$

where ρ is the density in kg/m^3 and c is the speed of sound in m/s. The SI unit for acoustic impedance is $kg/(m^2s)$, with the special name the *rayl*, where 1 rayl is equal to 1 $kg/(m^2s)$. Table 14-3 lists the acoustic impedances of materials and tissues commonly encountered in medical ultrasonography. In a simplistic way, the acoustic impedance can be likened to the stiffness and flexibility of a compressible medium such as a spring, as mentioned at the beginning of this chapter. When springs with different compressibility are connected together, the energy transfer from one spring to another depends mainly on stiffness. A large difference in the stiffness results in a large reflection of energy, an extreme example of which is a spring attached to a wall. Minor differences in stiffness or compressibility allow the continued propagation of energy, with little reflection at the interface. Sound propagating through a patient behaves similarly. Soft tissue adjacent to air-filled lungs represents a large difference in acoustic impedance; thus, ultrasonic energy incident on the lungs from soft tissue is almost entirely reflected. When adjacent tissues have similar acoustic impedances, only minor reflections of the incident energy occur. Acoustic impedance gives rise to differences in transmission and reflection of ultrasound energy, which is the means for producing an image using pulse-echo techniques.

TABLE 14-3 ACOUSTIC IMPEDANCE, $Z = \rho c$, FOR AIR, WATER, AND SELECTED TISSUES

TISSUE	Z (RAYLS)
Air	0.0004×10^6
Lung	0.18×10^6
Fat	1.34×10^6
Water	1.48×10^6
Kidney	1.63×10^6
Blood	1.65×10^6
Liver	1.65×10^6
Muscle	1.71×10^6
Skull bone	7.8×10^6

Reflection

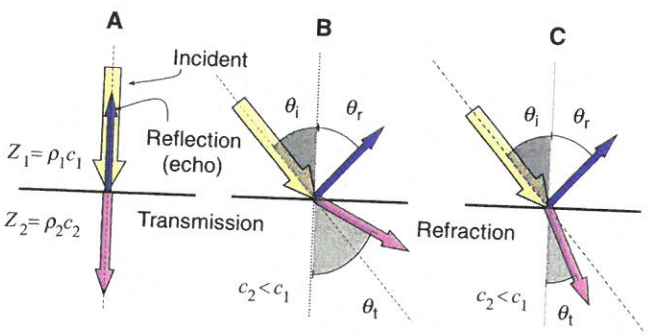
The reflection of ultrasound energy at a boundary between two tissues occurs because of the differences in the acoustic impedances of the two tissues. The reflection coefficient describes the fraction of sound intensity incident on an interface that is reflected. For perpendicular incidence (Fig. 14-5A), the *intensity* reflection coefficient, R_I, is expressed as

$$R_I = \frac{I_r}{I_i} = \left(\frac{Z_2 - Z_1}{Z_2 + Z_1}\right)^2$$

The subscripts 1 and 2 represent tissues proximal and distal to the boundary. The intensity transmission coefficient, T_I, is defined as the fraction of the incident intensity that is transmitted across an interface. With conservation of energy, the intensity transmission coefficient is $T_I = 1 - R_I$. For a fat-muscle interface, the intensity reflection and transmission coefficients are calculated as

$$R_{I,\,(Fat \rightarrow Muscle)} = \frac{I_r}{I_i} = \left(\frac{1.71 - 1.34}{1.71 + 1.34}\right)^2 = 0.015; \quad T_{I,\,(Fat \rightarrow Muscle)} = 1 - R_{I,\,(Fat \rightarrow Muscle)} = 0.985$$

The actual intensity reflected or transmitted at a boundary is the product of the incident intensity and the reflection coefficient. For example, an intensity of 40 mW/cm^2 incident on a boundary with R_I = 0.015 reflects 40 × 0.015 = 0.6 mW/cm^2. The intensity reflection coefficient at a tissue interface is readily calculated from the acoustic impedance of each tissue. Examples of tissue interfaces and respective reflection coefficients are listed in Table 14-4. For a typical muscle-fat interface, approximately 1% of the ultrasound intensity is reflected, and thus almost 99% of the intensity is transmitted to greater depths in the tissues. At a muscle-air interface, nearly 100% of incident intensity is reflected, making anatomy unobservable beyond an air-filled cavity. This is why acoustic coupling gel must be used between the face of the transducer and the skin to eliminate air pockets. A conduit of tissue that allows ultrasound transmission through structures such as the lung is known as an "acoustic window."



■ **FIGURE 14-5** Reflection and refraction of ultrasound occur at tissue boundaries with differences in acoustic impedance, Z. **A.** With perpendicular incidence (90 degrees), a fraction of the beam is transmitted and a fraction of the beam is reflected to the source at a tissue boundary. **B.** With nonperpendicular incidence, (90 degrees), the reflected fraction of the beam is directed away from the transducer at an angle $\theta_r = \theta_i$ and the transmitted fraction of the beam is refracted in the transmission medium at a transmitted refraction angle greater than the incident angle ($\theta_t > \theta_i$) when $c_2 > c_1$. **C.** Refraction angle of the transmitted beam is less than that of the incident angle when $c_1 > c_2$.

TABLE 14-4 PRESSURE AND REFLECTION COEFFICIENTS FOR VARIOUS INTERFACES

TISSUE INTERFACE	PRESSURE REFLECTION	INTENSITY REFLECTION
Liver-Kidney	−0.006	0.00003
Liver-Fat	−0.10	0.011
Fat-Muscle	0.12	0.015
Muscle-Bone	0.64	0.41
Muscle-Lung	−0.81	0.65
Muscle-Air	−0.99	0.99

When the beam is perpendicular to the tissue boundary, the sound is returned back to the transducer as an echo. As sound travels from a medium of lower acoustic impedance into a medium of higher acoustic impedance, the reflected wave experiences a 180-degree phase shift in pressure amplitude, as shown by the negative sign on the pressure amplitude values in Table 14-4. (Note: the reflected pressure amplitude is calculated as the ratio of the difference divided by the sum of the acoustic impedances.)

The above discussion assumes a "smooth" interface between tissues, where the wavelength of the ultrasound beam is much greater than the structural variations of the boundary. With higher frequency ultrasound beams, the wavelength becomes smaller, and the boundary no longer appears smooth. In this case, returning echoes are diffusely scattered throughout the medium, and only a small fraction of the incident intensity returns to the ultrasound transducer.

For nonperpendicular incidence at an angle θ_i (Fig. 14-5B,C), the ultrasound energy is reflected at an angle θ_r equal to the incident angle, $\theta_i = \theta_r$ Echoes are directed away from the source of ultrasound and are thus undetected.

Refraction

Refraction describes the change in direction of the transmitted ultrasound energy at a tissue boundary when the beam is not perpendicular to the boundary. Ultrasound frequency does not change when propagating through various media, but a change in the speed of sound may occur. Angles of incidence, reflection, and transmission are measured relative to the normal incidence on the boundary, as shown in Figure 14-5B,C. The angle of refraction (θ_t) is determined by the speed of sound change that occurs at the boundary and is related to the angle of incidence (θ_i) by Snell's law:

$$\frac{\sin\theta_t}{\sin\theta_i} = \frac{c_2}{c_1}$$

where θ_i and θ_t are the incident and transmitted angles, c_1 and c_2 are the speeds of sound in medium 1 and 2, and medium 2 carries the transmitted ultrasound energy. For small angles of incidence and transmission, Snell's law can be approximated as

$$\frac{\theta_t}{\theta_i} \cong \frac{c_2}{c_1}$$

When $c_2 > c_1$, the angle of transmission is greater than the angle of incidence as shown in Figure 14-5B, and the opposite when $c_2 < c_1$, as shown in Figure 14-5C. No refraction

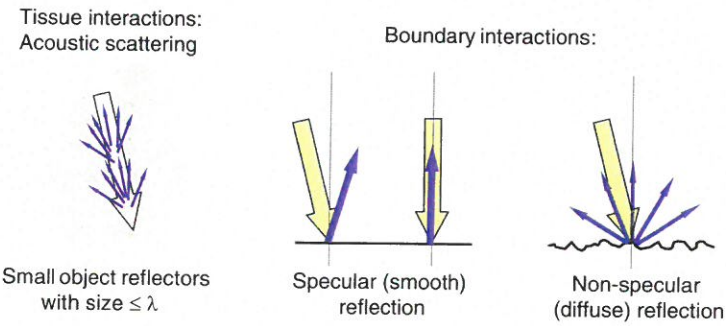
occurs when the speed of sound is the same in the two media, or with perpendicular incidence. This straight-line propagation is assumed in ultrasound signal processing, and when refraction does occur, misplacement of anatomy in the image can result.

A situation called total reflection occurs when $c_2 > c_1$ and the angle of incidence of the sound beam with the boundary between two media exceeds an angle called the critical angle. In this case, the sound beam does not penetrate the second medium at all but travels along the boundary. The critical angle (θ_c) is calculated by setting θ_t = 90 degree in Snell's law (equation above), producing the equation $\sin\theta_c = c_1/c_2$.

Scattering

Acoustic scattering arises from objects and interfaces within a tissue that are about the size of the wavelength or smaller and represent a rough or *nonspecular reflector* surface, as shown in Figure 14-6. Most organs have a characteristic structure that gives rise to a defined scatter "signature" and provides much of the diagnostic information contained in the ultrasound image. A *specular reflector* is a smooth boundary between two media, where the dimensions of the boundary are much larger than the wavelength of the incident ultrasound energy. Because nonspecular reflectors reflect sound in all directions, the amplitudes of the returning echoes are significantly weaker than echoes from tissue boundaries. Fortunately, the dynamic range of the ultrasound receiver is sufficient to detect echo information over a wide range of amplitudes. Intensities of returning echoes from nonspecular reflectors in the tissue parenchyma are not greatly affected by beam direction, unlike the strong directional dependence of specular reflectors. Parenchyma-generated echoes typically have similar echo strengths and gray-scale levels in the image. Differences in scatter amplitude that occur from one region to another cause corresponding brightness changes on the ultrasound display.

In general, the echo signal amplitude from the insonated tissues depends on the number of scatterers per unit volume, the acoustic impedance differences at the scatterer interfaces, the sizes of the scatterers, and the ultrasonic frequency. Hyperechoic (higher scatter amplitude) and hypoechoic (lower scatter amplitude) are terms used for describing the scatter characteristics relative to the average background signal. Hyperechoic areas usually have greater numbers of scatterers, larger acoustic impedance differences, and larger scatterers. Acoustic scattering from nonspecular reflectors increases with frequency, while specular reflection is relatively independent



■ **FIGURE 14-6** Ultrasound scattering. Small particle reflectors within a tissue or organ cause a diffuse scattering pattern that is characteristic of the particle size, giving rise to specific tissue or organ "signatures." Boundary interactions with ultrasound can also scatter the beam, particularly at higher frequencies. Specular and nonspecular boundary characteristics are partially dependent on the wavelength of the incident ultrasound. As the wavelength becomes smaller, the boundary becomes "rough," resulting in scattering from the surface because of nonperpendicular boundaries.

consequence of frequency-dependent attenuation is the preferential removal of the highest frequency components in a broadband ultrasound pulse (discussed below in Section 14.3) and a shift to lower frequencies.

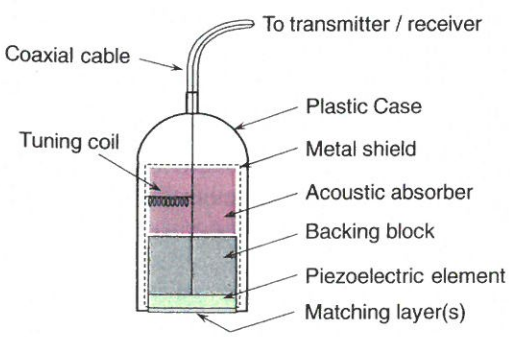
14.3 Ultrasound Transducers

Ultrasound is produced and detected with a *transducer*, comprised of one or more ceramic elements with electromechanical properties and peripheral components. The ceramic element converts electrical energy into mechanical energy to produce ultrasound and mechanical energy into electrical energy for ultrasound detection. Over the past several decades, the transducer assembly has evolved considerably in design, function, and capability, from a single-element resonance crystal to a broadband transducer array of hundreds of individual elements. A simple single-element, plane-piston source transducer is illustrated in Figure 14-8. Major components include the piezoelectric material, matching layer, backing block, acoustic absorber, insulating cover, tuning coil, sensor electrodes, and transducer housing.

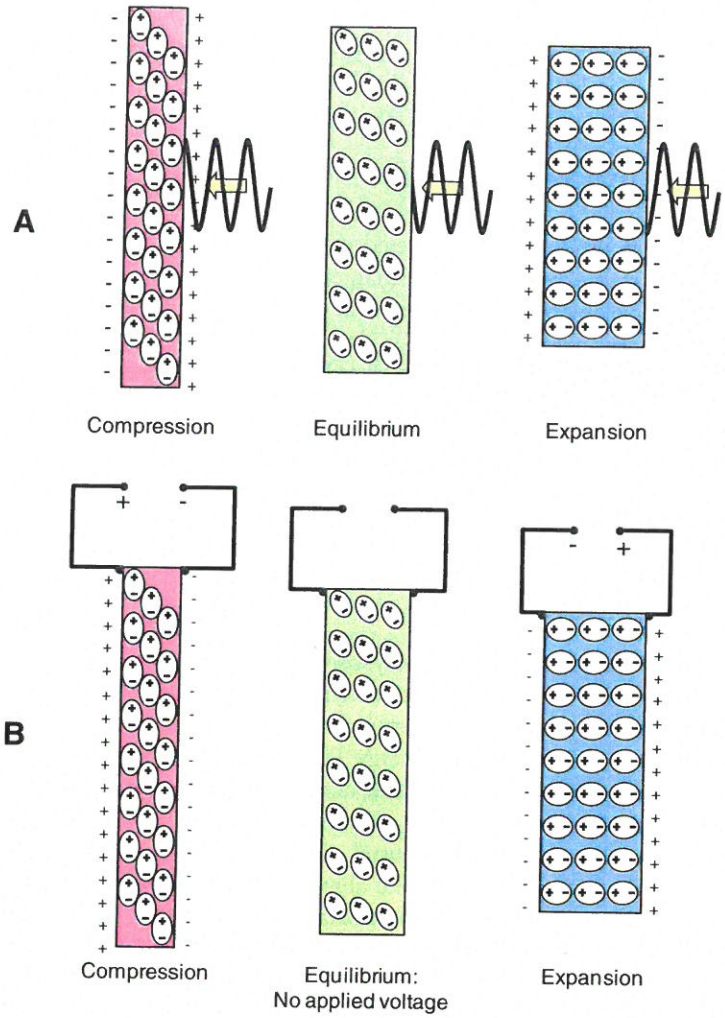
Piezoelectric Materials

A piezoelectric material (often a crystal or ceramic) is the functional component of the transducer. It converts electrical energy into mechanical (sound) energy by physical deformation of the crystal structure. Conversely, mechanical pressure applied to its surface creates electrical energy. Piezoelectric materials are characterized by a well-defined molecular arrangement of electrical dipoles, as shown in Figure 14-9. Electrical dipoles are molecular entities containing positive and negative electric charges that have an overall neutral charge. When mechanically compressed by an externally applied pressure, the alignment of the dipoles is disturbed from the equilibrium position to cause an imbalance of the charge distribution. A potential difference (voltage) is created across the element with one surface maintaining a net positive charge and one surface a net negative charge. Surface electrodes measure the magnitude of voltage, which is proportional to the incident mechanical pressure amplitude. Conversely, application of an external voltage through conductors attached to the surface electrodes induces the mechanical expansion and contraction of the transducer element.

Piezoelectric materials are available from natural and synthetic materials. An example of a natural piezoelectric material is quartz crystal, commonly used in watches and other timepieces to provide a mechanical vibration source at 32.768 kHz for interval timing. (This is one of several oscillation frequencies of quartz, determined by the

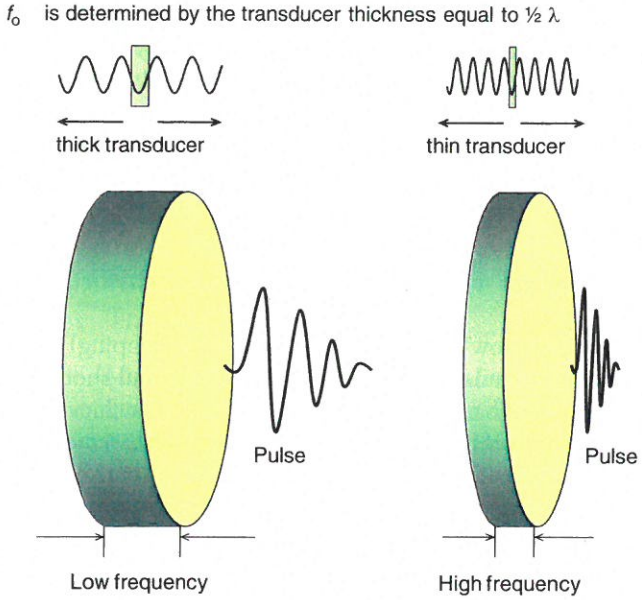


■ **FIGURE 14-8** A single-element ultrasound transducer assembly is comprised of the piezoelectric ceramic, the backing block, acoustic absorber, tuning coil and metal shield, transducer housing, coaxial cable and voltage source, and the ceramic to tissue matching layer.



■ FIGURE 14-9 The piezoelectric element is comprised of aligned molecular dipoles. **A.** Under the influence of mechanical pressure from an adjacent medium (e.g., an ultrasound echo), the element thickness contracts (at the peak pressure amplitude), achieves equilibrium (with no pressure), or expands (at the peak rarefactional pressure) causing realignment of the electrical dipoles to produce positive and negative surface charge. Surface electrodes (not shown) measure the charge variation (voltage) as a function of time **B.** An external voltage source applied to the element surfaces causes compression or expansion from equilibrium by realignment of the dipoles in response to the electrical attraction or repulsion force.

crystal cut and machining properties.) Ultrasound transducers for medical imaging applications employ a synthetic piezoelectric ceramic, most often *lead-zirconate-titanate* (PZT)—a compound with the structure of molecular dipoles. The piezoelectric attributes are attained after a process of molecular synthesis, heating, orientation of internal dipole structures with an applied external voltage, cooling to permanently maintain the dipole orientation, and cutting into a specific shape. For PZT in its natural state, no piezoelectric properties are exhibited; however, heating the material past its "Curie temperature" (e.g., 328°C to 365°C) and applying an external voltage causes the dipoles to align in the ceramic. The external voltage is maintained until the material has cooled to below its Curie temperature. Once the material has cooled, the dipoles retain their alignment. At equilibrium, there is no net charge on ceramic



■ **FIGURE 14-10** A short-duration voltage spike causes the resonance piezoelectric element to vibrate at its natural frequency, f_0, which is determined by the thickness of the transducer equal to ½ λ. Low-frequency oscillation is produced with a thicker piezoelectric element. The spatial pulse length (SPL) is a function of the operating frequency and the adjacent damping block.

surfaces. When compressed, an imbalance of charge produces a voltage between the surfaces. Similarly, when a voltage is applied between electrodes attached to both surfaces, mechanical deformation occurs, as illustrated in Figure 14-9A,B.

Resonance Transducers

Resonance transducers for pulse-echo ultrasound imaging operate in a "resonance" mode, whereby a voltage (usually 150 V) of very short duration (a voltage spike of ~1 μs) is applied, causing the piezoelectric material to initially contract and then subsequently vibrate at a natural resonance frequency. This frequency is selected by the "thickness cut" of the PZT, due to the preferential emission of ½-wavelength ultrasound waves in the piezoelectric material as illustrated in Figure 14-10. The operating frequency is determined from the speed of sound in, and the thickness of, the piezoelectric material. For example, a 5-MHz transducer will have a wavelength in PZT (speed of sound in PZT is ~4,000 *m/s*) of

$$\lambda = \frac{c}{f} = \frac{4{,}000\,m/s}{5\times10^{6}/s} = 8 \times 10^{4}\,m = 0.8\,mm$$

To achieve the 5-MHz resonance frequency, a transducer element thickness of ½ × 0.8 *mm* = 0.4 *mm* is required. Higher frequencies are achieved with thinner elements, and lower frequencies with thicker elements. Resonance transducers transmit and receive preferentially at a single "center frequency."

Damping Block

The damping block, layered on the back of the piezoelectric element, absorbs the backward directed ultrasound energy and attenuates stray ultrasound signals from the housing. This component also dampens the transducer vibration to create an ultrasound pulse with a short spatial pulse length (SPL), which is necessary to preserve detail along the beam axis (axial resolution). Dampening of the vibration

(also known as "ring-down") lessens the purity of the resonance frequency and introduces a broadband frequency spectrum. With ring-down, an increase in the bandwidth (range of frequencies) of the ultrasound pulse occurs by introducing higher and lower frequencies above and below the center (resonance) frequency. The "Q factor" describes the bandwidth of the sound emanating from a transducer as

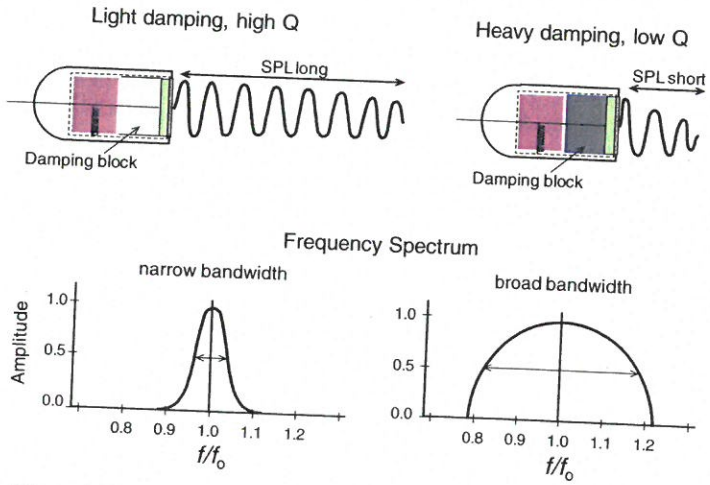
$$Q = \frac{f_0}{\text{bandwidth}}$$

where f_0 is the center frequency and the bandwidth is the width of the frequency distribution.

A "high Q" transducer has a narrow bandwidth (i.e., very little damping) and a corresponding long SPL. A "low Q" transducer has a wide bandwidth and short SPL. Imaging applications require a broad bandwidth transducer in order to achieve high spatial resolution along the direction of beam travel. Blood velocity measurements by Doppler instrumentation (explained in Section 14.9) require a relatively narrow-band transducer response in order to preserve velocity information encoded by changes in the echo frequency relative to the incident frequency. Continuous-wave ultrasound transducers have a very high Q characteristic. An example of a "high Q" and "low Q" ultrasound pulse illustrates the relationship to SPL in Figure 14-11. While the Q factor is derived from the term "quality factor," a transducer with a low Q does not imply poor quality in the signal.

Matching Layer

The *matching layer* provides the interface between the raw transducer element and the tissue and minimizes the acoustic impedance differences between the transducer and the patient. It consists of layers of materials with acoustic impedances that are intermediate to soft tissue and the transducer material. The thickness of each layer is equal to ¼ wavelength, determined from the center operating frequency of the



■ **FIGURE 14-11** Effect of damping block on the frequency spectrum. The damping block is adjacent to the backside of the transducer and limits the vibration of the element to a small number of cycles. Light damping allows many cycles to occur (**top left**), which results in an extended SPL (number of cycles times the wavelength) and a narrow frequency bandwidth (range of frequencies contained in the pulse (**bottom left**). Heavy damping reduces the SPL and broadens the frequency bandwidth (**top and bottom right**). The Q factor describes the center frequency divided by the bandwidth.

(the ability of the system to resolve objects in a direction perpendicular to the beam direction) is dependent on the beam diameter and is best at the end of the near field for a single-element transducer. Lateral resolution is poor in areas close to and far from the transducer surface.

Pressure amplitude characteristics in the near field are very complex, caused by the constructive and destructive interference wave patterns of the ultrasound beam. Peak ultrasound pressure occurs at the end of the near field, corresponding to the minimum beam diameter for a single-element transducer. Pressures vary rapidly from peak compression to peak rarefaction several times during transit through the near field. Only when the far field is reached do the ultrasound pressure variations decrease continuously (Fig. 14-14B).

The Far Field

The far field is also known as the Fraunhofer zone and is where the beam diverges. For a large-area single-element transducer, the angle of ultrasound beam divergence, θ, for the far field is given by

$$\sin\theta = 1.22\frac{\lambda}{d}$$

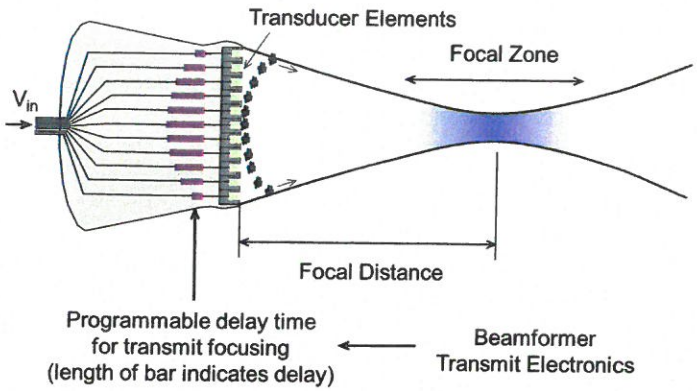
where d is the effective diameter of the transducer and λ is the wavelength. Less beam divergence occurs with high-frequency, large-diameter transducers. Unlike the near field, where beam intensity varies from maximum to minimum to maximum in a converging beam, ultrasound intensity in the far field decreases monotonically with distance.

Transducer Array Beam Formation and Focusing

In a transducer array, the narrow piezoelectric element width (typically between one-half to one wavelength) produces a diverging beam at a distance very close to the transducer face. Formation and convergence of the ultrasound beam occur with the operation of several or all of the transducer elements at the same time. Transducer elements in a linear array that are fired simultaneously produce an effective transducer width equal to the sum of the widths of the individual elements. Individual beams interact via "constructive" and "destructive" interference to produce a collimated beam that has properties similar to the properties of a single transducer of the same size. With a phased-array transducer, the beam is formed by interaction of the individual wave fronts from each transducer, each with a slight difference in excitation time. Minor phase differences of adjacent beams form constructive and destructive wave summations that "steer" or "focus" the beam profile.

Transmit Focus

For a single transducer or group of simultaneously fired transducers in a linear array, the focal distance is a function of the transducer diameter, the center operating frequency, and the presence of any acoustic lenses attached to the element surface. This focal depth is unchangeable. Phased array transducers and many linear array transducers allow a selectable focal distance by applying specific timing delays between transducer elements that cause the beam to converge at a specified distance. A shallow focal zone (close to the transducer surface) is produced by firing outer transducers in the array before the inner transducers in a symmetrical pattern, as shown in Figure 14-15. Greater focal distances are achieved by reducing the delay time differences amongst the transducer elements, resulting in more distal beam convergence. Multiple



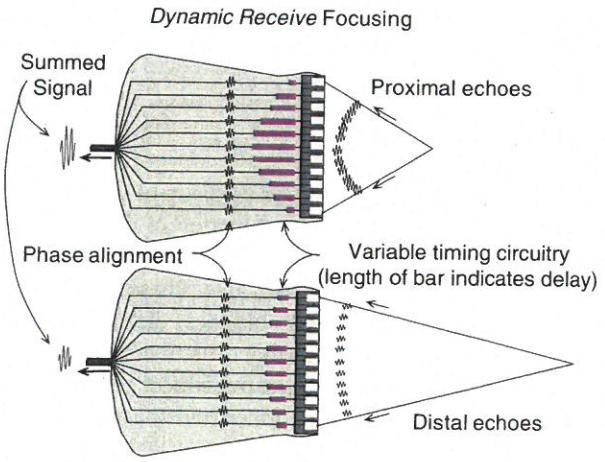
■ **FIGURE 14-15** A phased-array transducer assembly uses all elements to produce the ultrasound beam. Focusing is achieved by implementing a programmable delay time (beam former electronics) for the excitation of the individual transducer elements (focusing requires the outer elements in the array be energized first). Phase differences of the individual ultrasound pulses result in a minimum beam diameter (the focal distance) at a predictable depth in tissue.

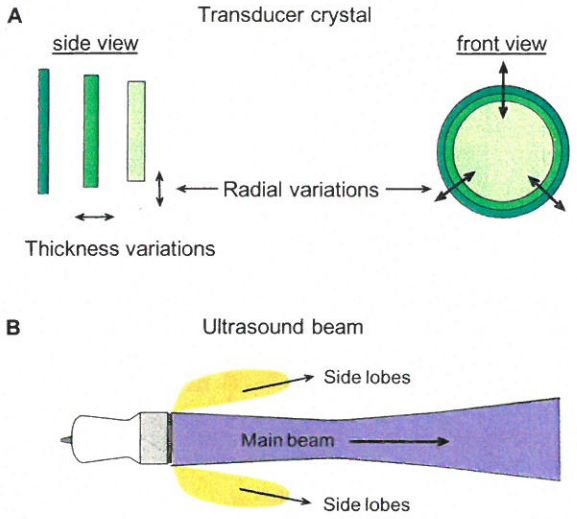
transmit focal zones are created by repeatedly acquiring data over the same volume, but with different phase timing of the transducer array elements.

Receive Focus

In a phased-array transducer, the echoes received by all of the individual transducer elements are summed together to create the ultrasound signal from a given depth. Echoes received at the edge of the element array travel a slightly longer distance than those received at the center of the array, particularly at shallow depths. Signals from individual transducer elements therefore must be rephased to avoid a loss of resolution when the individual signals are synthesized to make an image. *Dynamic receive focusing* is a method to rephase the signals by dynamically introducing electronic delays as function of depth (time). At shallow depths, rephasing delays between adjacent transducer elements are greatest. With greater depth, there is less difference, so the phase delay circuitry for the receiver varies as a function of echo listening time, as shown in Figure 14-16. In addition to phased-array transducers, many linear array transducers permit dynamic receive focusing amongst the active element group.

■ **FIGURE 14-16** Dynamic receive focusing. All transducer elements in the phased array are active during the receive mode, and to maintain focus, the receive focus timing must be continuously adjusted in order to compensate for differences in arrival time across the array as a function of time (depth of the echo). Depicted are an early time (top illustration) of proximal echo arrival and a later time of distal echo arrival. To achieve phase alignment of the echo responses by all elements, variable timing is implemented as a function of element position after the transmit pulse in the beam former. The output of all phase-aligned echoes is summed.





■ **FIGURE 14-17** **A.** A single transducer of a multielement array produces the main ultrasound beam in "thickness mode" vibration; however, radial expansion and contraction also occur. **B.** Side lobe energy in a multielement transducer array is created from the radial transducer vibration, which is directed away from the main beam. Echoes received from the side lobe energy are mapped into the main beam, creating unwanted artifacts.

Dynamic Aperture

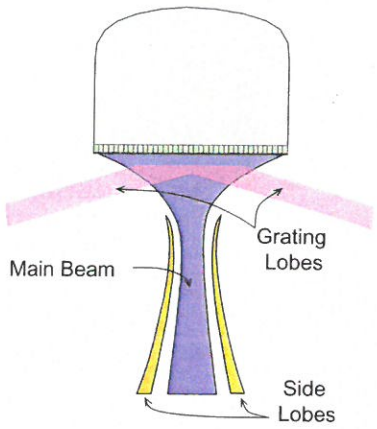
The lateral spatial resolution of the linear array beam varies with depth, dependent on the linear dimension of the transducer width (aperture). A process termed *dynamic aperture* increases the number of active *receiving* elements in the array with reflector depth, so that the lateral resolution does not degrade with depth of propagation.

Side Lobes and Grating Lobes

Side lobes are unwanted emissions of ultrasound energy directed away from the main pulse, caused by the radial expansion and contraction of the transducer element during thickness contraction and expansion (Fig. 14-17). In the receive mode of transducer operation, echoes generated from the side lobes are unavoidably remapped along the main beam, which can introduce artifacts in the image. In continuous mode operation, the narrow frequency bandwidth of the transducer (high Q) causes the side lobe energy to be a significant fraction of the total beam. In pulsed mode operation, the low Q, broadband ultrasound beam produces a spectrum of acoustic wavelengths that reduce the emission of side lobe energy.

For multielement arrays, side lobe emission occurs in a forward direction along the main beam. By keeping the individual transducer element widths small (less than ½ wavelength), the side lobe emissions are reduced. Another method to minimize side lobes with array transducers is to reduce the amplitude of the peripheral transducer element excitations relative to the central element excitations.

Grating lobes result when ultrasound energy is emitted far off-axis by multielement arrays and are a consequence of the noncontinuous transducer surface of the discrete elements. The grating lobe effect is equivalent to placing a grating in front of a continuous transducer element, producing coherent waves directed at a large angle away from the main beam (Fig. 14-18). This misdirected energy of relatively low amplitude can result in the appearance of highly reflective objects in the main beam.



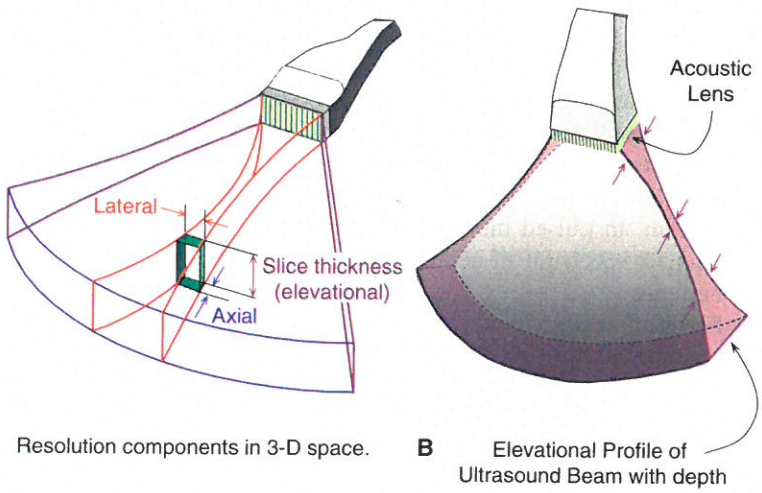
■ **FIGURE 14-18** Side and grating lobes are off-axis energy emissions produced by linear and phased-array transducers. Side lobes are forward directed; grating lobes are emitted from the array surface at very large angles.

Spatial Resolution

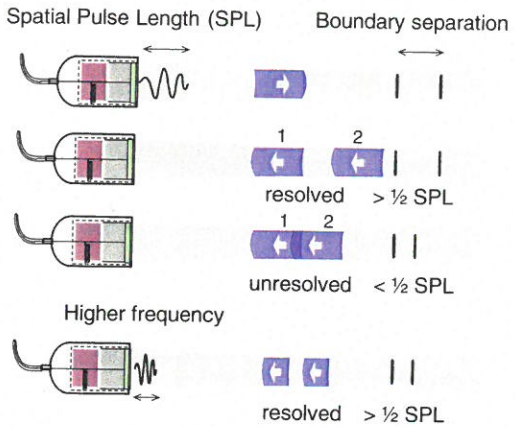
In ultrasound, the major factor that limits the spatial resolution and visibility of detail is the volume of the acoustic pulse. The axial, lateral, and elevational (slice-thickness) dimensions determine the minimal volume element (Fig. 14-19). Each dimension has an effect on the resolvability of objects in the image.

Axial Resolution

Axial resolution (also known as linear, range, longitudinal, or depth resolution) refers to the ability to discern two closely spaced objects in the direction of the beam. Achieving good axial resolution requires that the returning echoes be distinct without overlap. The minimal required separation distance between two reflectors is



■ **FIGURE 14-19** **A.** The axial, lateral, and elevational (slice-thickness) contributions in three dimensions are shown for a phased-array transducer ultrasound beam. Axial resolution, along the direction of the beam, is independent of depth; lateral resolution and elevational resolution are strongly depth dependent. Lateral resolution is determined by transmit and receive focus electronics; elevational resolution is determined by the height of the transducer elements. At the focal distance, axial is better than lateral and lateral is better than elevational resolution. **B.** Elevational resolution profile with an acoustic lens across the transducer array produces a focal zone in the slice-thickness direction.



■ **FIGURE 14-20** Axial resolution is equal to ½ SPL. Tissue boundaries that are separated by a distance greater than ½ SPL produce echoes from the first boundary that are completely distinct from echoes reflected from the second boundary, whereas boundaries with less than ½ SPL result in overlap of the returning echoes. Higher frequencies reduce the SPL and thus improve the axial resolution, as shown in the lower diagram.

one-half of the SPL to avoid the overlap of returning echoes, as the distance traveled between two reflectors is twice the separation distance. Objects spaced closer than ½ SPL will not be resolved (Fig. 14-20).

The SPL is the number of cycles emitted per pulse by the transducer multiplied by the wavelength. Shorter pulses, producing better axial resolution, can be achieved with greater damping of the transducer element (to reduce the pulse duration and number of cycles) or with higher frequency (to reduce wavelength). For imaging applications, the ultrasound pulse typically consists of three cycles. At 5 MHz (wavelength of 0.31 *mm*), the SPL is about $3 \times 0.31 = 0.93$ *mm*, which provides an axial resolution of ½ (0.93 *mm*) = 0.47 *mm*. At a given frequency, shorter pulse lengths require heavy damping and low Q, broad bandwidth operation. For a constant damping factor, higher frequencies (shorter wavelengths) give better axial resolution, but the imaging depth is reduced. Axial resolution remains constant with depth.

Lateral Resolution

Lateral resolution, also known as azimuthal resolution, refers to the ability to discern as separate two closely spaced objects perpendicular to the beam direction. For both single-element transducers and multielement array transducers, the beam diameter determines the lateral resolution (see Fig. 14-21). Since the beam diameter varies with distance from the transducer in the near and far field, the lateral resolution is depth dependent. The best lateral resolution occurs at the near field–far field interface. At this depth, the effective beam diameter is approximately equal to ½ the transducer diameter. In the far field, the beam diverges and substantially reduces the lateral resolution.

The lateral resolution of linear and curvilinear array transducers can be varied. The number of elements simultaneously activated in a group defines an "effective" transducer width that has similar behavior to a single-transducer element of the same width. Transmit and receive focusing along a single line is possible to produce focal zones with depth along each line.

For the phased-array transducer, focusing to a specific depth is achieved by both beam steering and transmit/receive focusing to reduce the effective beam width and improve lateral resolution, especially in the near field. Multiple transmit/receive focal zones can be implemented to maintain lateral resolution as a function of depth (Fig. 14-22). Each focal zone requires separate pulse-echo sequences to acquire data. One way to accomplish this is to acquire data along one beam direction multiple

2. **Dynamic range (logarithmic) compression.** *Dynamic range* defines the effective operational range of an electronic device from the threshold signal level to the saturation level. Key components in the ultrasound detection and display that are most affected by a wide dynamic range include the ADC and the display. For receiver systems with an 8-bit ADC, the dynamic range is 20 log (256) = 48 dB, certainly not enough to accurately digitize a 50-dB signal. Video monitors and film have a dynamic range of about 100:1 to 150:1 and therefore require a reduced range of input signals to accurately render the display output. Thus, after TGC, the signals must be reduced to 20 to 30 dB, which is accomplished by compression using *logarithmic amplification* to increase the smallest echo amplitudes and to decrease the largest amplitudes. Logarithmic amplification produces an output signal proportional to the logarithm of the input signal. Logarithmic amplification is performed by an analog signal processor in less costly ultrasound systems and digitally in high-end digital systems having ADCs with a large bit depth. In any case, an appropriate range of signals is achieved for display of the amplitude variations as gray scale on the monitor or for printing on film.
3. **Rectification, demodulation, and envelope detection.** *Rectification* inverts the negative amplitude signals of the echo to positive values. *Demodulation and envelope detection* convert the rectified amplitudes of the echo into a smoothed, single pulse.
4. **Rejection** level adjustment sets the threshold of signal amplitudes allowed to pass to the digitization and display subsystems. This removes a significant amount of undesirable low-level noise and clutter generated from scattered sound or by the electronics.
5. **Processed images** are optimized for gray-scale range and viewing on the limited dynamic range monitors, so that subsequent adjustments to the images are unnecessary.

In the above listed steps, the operator has the ability to control the TGC and noise/clutter rejection. The amount of amplification (overall gain) necessary is dependent on the initial power (transmit gain) settings of the ultrasound system. Higher intensities are achieved by exciting the transducer elements with larger voltages. This increases the amplitude of the returning echoes and reduces the need for electronic amplification gain but also deposits more energy into the patient, where heating or mechanical interactions can be significant. Conversely, lower ultrasound power settings, such as those used in obstetrical ultrasound, require a greater overall electronic gain to amplify weaker echo signals into the appropriate range for TGC. TGC allows the operator to manipulate depth-dependent gain to improve image uniformity and compensate for unusual imaging situations. Inappropriate adjustment of TGC can lead to artifactual enhancement of tissue boundaries and tissue texture as well as nonuniform response versus depth. The noise rejection level sets a threshold to clean up low-level signals in the electronic signal. It is usually adjusted in conjunction with the transmit power level setting of the ultrasound instrument.

Echo Display Modes

A-mode

A-mode (A for amplitude) is the display of the *processed* information from the receiver versus time (after the receiver processing steps shown in Fig. 14-27). As echoes return from tissue boundaries and scatterers (a function of the acoustic impedance differences in the tissues), a digital signal proportional to echo amplitude is produced

as a function of time. One "A-line" of data per PRP is the result. As the speed of sound equates to depth (round-trip time), the tissue interfaces along the path of the ultrasound beam are localized by distance from the transducer. The earliest uses of ultrasound in medicine used A-mode information to determine the midline position of the brain for revealing possible mass effect of brain tumors. A-mode and A-line information is currently used in ophthalmology applications for precise distance measurements of the eye. Otherwise, A-mode display by itself is seldom used.

B-mode

B-mode (B for brightness) is the electronic conversion of the A-mode and A-line information into brightness-modulated dots along the A-line trajectory. In general, the brightness of the dot is proportional to the echo signal amplitude (depending upon signal processing parameters). The B-mode display is used for M-mode and 2D gray-scale imaging.

M-mode

M-mode (M for motion) is a technique that uses B-mode information to display the echoes from a moving organ, such as the myocardium and valve leaflets, from a fixed transducer position and beam direction on the patient (Fig. 14-29). The echo data from a single ultrasound beam passing through moving anatomy are acquired and displayed as a function of time, represented by reflector depth on the vertical axis (beam path direction) and time on the horizontal axis. M-mode can provide excellent temporal resolution of motion patterns, allowing the evaluation of the function of heart valves and other cardiac anatomy. Only one anatomical dimension is represented by the M-mode technique, and with advances in real-time 2D echocardiography, Doppler, and color flow imaging, this display mode is of much less importance than in the past.

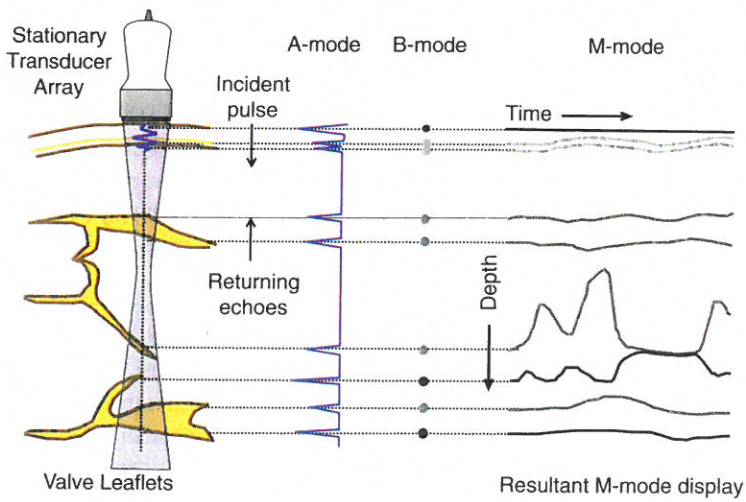


FIGURE 14-29 Ultrasonic data are acquired with a stationary transducer or transducer array. The A-mode data (vertical ***blue*** trace) represent echo amplitudes during one pulse-echo period. The amplitudes are encoded to brightness (B-mode) as a series of variable intensity dots. The A-mode line and corresponding B-mode dots change position as the valve leaflets move within the stationary beam. By deflecting the dots horizontally in time on a storage display, motion graphs (M-mode) are created, depicting the periodic (or lack of periodic) motion.

Image Storage

Ultrasound images are typically comprised of 640 × 480 or 512 × 512 pixels. Each pixel has a depth of 8 bits (1 byte) of digital data, providing up to 256 levels of gray scale. Image storage (without compression) is approximately ¼ MB per image. For real-time imaging (10 to 30 frames per second), this can amount to hundreds of megabytes of data for video clip acquisitions, depending on the complexity of the examination. Color images used for Doppler studies (Section 14.7) increase the storage requirements further because of larger numbers of bits needed for color resolution (full fidelity color requires 24 bits/pixel, one byte each for the red, green, and blue primary colors).

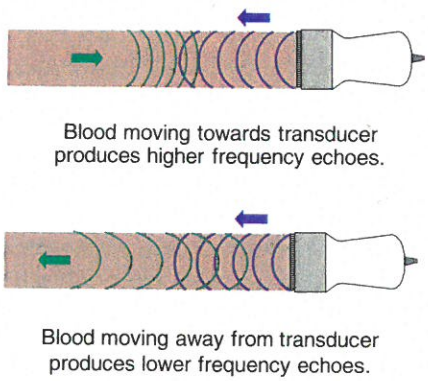
14.7 Doppler Ultrasound

Doppler ultrasound is based on the shift of frequency in an ultrasound wave caused by a moving reflector, such as blood cells in the vasculature (Fig. 14-36). This is the same effect that causes a siren on a fire truck to sound high pitched as the truck approaches the listener (the wavelength is compressed) and a shift to a lower pitch sound as it passes by and continues on (the wavelength is expanded). The moving reflectors in the body are the blood cells. By comparing the incident ultrasound frequency with the reflected ultrasound frequency from the blood cells, it is possible to discern the velocity of the blood. Not only can blood velocity (and indirectly blood flow) be measured, but the information provided by the Doppler techniques can also be used to create color blood flow maps of the vasculature. The interpretation of Doppler signals in clinical practice, however, requires the extraction of information about the blood flow from the potential confounding aspects related to the technique itself. Therefore, an understanding of the physical principles of Doppler ultrasound is an absolute prerequisite for the interpretation of the acquired information.

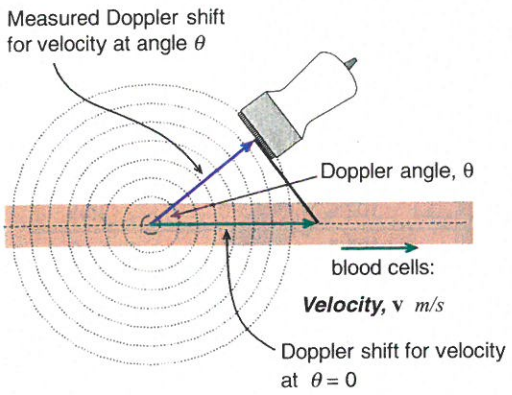
Doppler Frequency Shift

The *Doppler shift* is the difference between the incident frequency and reflected frequency. When the reflector is moving directly away from or toward the source of sound, the Doppler frequency shift (f_d) is calculated as

$$f_d = f_r - f_i = \frac{\text{reflector speed}}{\text{reflector speed + speed of sound}} \times 2 \times f_i$$



■ **FIGURE 14-36** Doppler ultrasound exploits changes in frequency from interaction with moving objects. Sound waves reflected from a moving object are compressed (higher frequency) when moving toward the transducer and expanded (lower frequency) when moving away from the transducer compared to the incident sound wave frequency. The difference between the incident and returning frequencies is called the Doppler shift frequency.



■ **FIGURE 14-37** Geometry of Doppler ultrasound data acquisition. The Doppler shift varies as a function of the angle (θ) of the incident ultrasound pulse and the axis of the blood vessel for a fixed blood velocity, v. The maximum Doppler shift occurs at an angle θ = 0 For a larger Doppler angle (θ), the measured Doppler shift is less by a factor of cos(θ), and velocity estimates are compensated by 1/cos (θ).

where f_i is the frequency of the sound incident on the reflector and f_r is the frequency of the reflected sound. Thus, the Doppler shift is proportional to the velocity of the blood cells.

When the sound waves and blood cells are not moving in parallel directions, the equation must be modified to account for less Doppler shift. The angle between the direction of blood flow and the direction of the sound is called the Doppler angle (Fig. 14-37). The component of the velocity vector directed toward the transducer is less than the velocity vector along the vessel axis by the cosine of the angle, cos (θ). Without correction for this discrepancy, the Doppler shift will be less and an underestimate of the actual blood velocity will occur. The Doppler angle joins the adjacent side and hypotenuse of a right triangle; therefore, the component of the blood velocity in the direction of the sound (adjacent side) is equal to the actual blood velocity (hypotenuse) multiplied by the cosine of the angle (θ). (Note: In physics, the term *velocity* refers to a vector quantity, describing both the distance traveled per unit time (*speed*) and the direction of movement such as blood flow.) As the velocity of blood cells (peak of ~200 *cm/s*) is significantly less than the speed of sound (154,000 *cm/s*), the denominator can be simplified with an extremely small error by neglecting the velocity of the blood. This results in the generalized Doppler shift equation:

$$f_d = \frac{2 f_i \, \mathrm{v} \cos(\theta)}{c}$$

where **v** is the velocity of blood, c is the speed of sound in soft tissue, and θ is the Doppler angle.

Calculation of the blood velocity is straightforward by rearranging the Doppler equation:

$$\mathrm{v} = \frac{f_d c}{2 f_i \cos(\theta)}$$

Thus, the measured Doppler shift at a Doppler angle θ is adjusted by 1/cos(θ) in order to achieve accurate velocity estimates. Selected cosine values are cos 0° = 1, cos 30° = 0.87, cos 45° = 0.707, cos 60° = 0.5, and cos 90° = 0. At a 60-degree Doppler angle, the measured Doppler frequency is ½ the actual Doppler frequency, and at 90 degrees, the measured frequency is 0. The *preferred* Doppler angle ranges from 30 to 60 degrees. At too large an angle (greater than 60 degrees), the apparent

Artifacts

Artifacts arise from the incorrect display of anatomy or noise during imaging. The causes are machine and operator related, as well as intrinsic to the characteristics and interaction of ultrasound with the tissues. Understanding how artifacts are generated in ultrasound is crucial, which places high demands on the knowledge and experience of the sonographer.

Artifacts can be caused by a variety of mechanisms. For instance, sound travels at different speeds, not just the 1,540 *m/s* average value for soft tissue that is used in the "range equation" for placing echo information in the image matrix. This speed variation results in some echoes being displaced from the expected location in the image display. Sound is refracted when the beam is not perpendicular to a tissue boundary; echoes are deflected to the receiver from areas outside of the main beam and can be mismapped into the image. Improper use of TGC causes suboptimal image quality with nonuniform appearance of tissues within a given band of the image.

Fortunately, most ultrasound artifacts are discernible to the experienced sonographer because of transient appearance and/or obvious effects on the image. Some artifacts are used to advantage as a diagnostic aid in characterization of tissue structures and their composition. Besides aliasing in Doppler and color flow acquisitions discussed in Section 14.7, many common artifacts in B-mode imaging are discussed below.

Refraction

Refraction is a change in the transmitted ultrasound pulse direction at a boundary with nonperpendicular incidence, when the two tissues support a different speed of sound ($C_1 \neq C_2$). Misplaced anatomy often occurs in the image during the scan (Fig. 14-52A). The ultrasonographer must be aware of objects appearing and disappearing with slight differences in orientation of the beam. Anatomical displacement due to refraction artifacts will change with the position of the transducer and angle of incidence with the tissue boundaries.

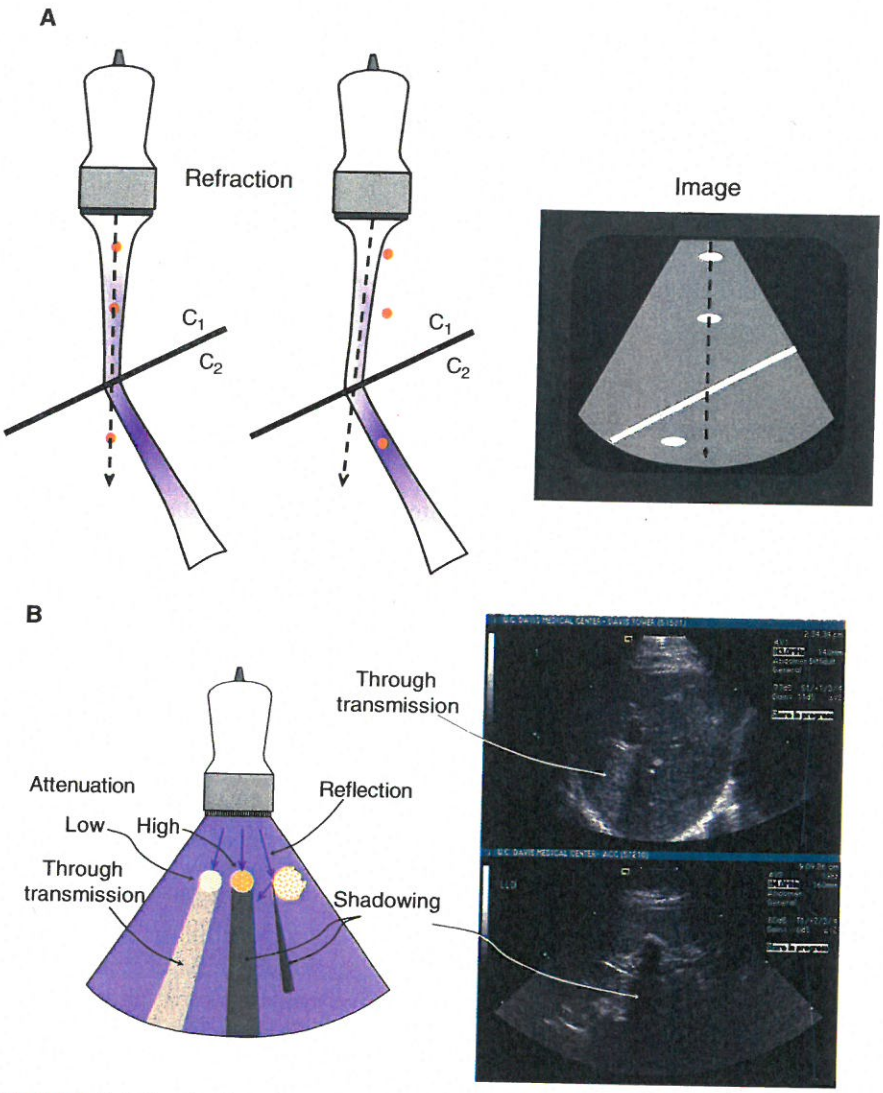
Shadowing and Enhancement

Shadowing is a hypointense signal area distal to an object or interface and is caused by objects with high attenuation or reflection of the incident beam without the return of echoes. Highly attenuating objects such as bones or kidney stones reduce the intensity of the transmitted beam and can induce low-intensity streaks in the image. Reflection of the incident beam from curved surfaces eliminates the distal propagation of ultrasound and causes streaks or shadowing. Enhancement occurs distal to objects having very low ultrasound attenuation, such as fluid-filled cavities (e.g., a filled bladder or cysts). Hyperintense signals ("through transmission") arise from increased transmission of sound by these structures (Fig. 14-52B).

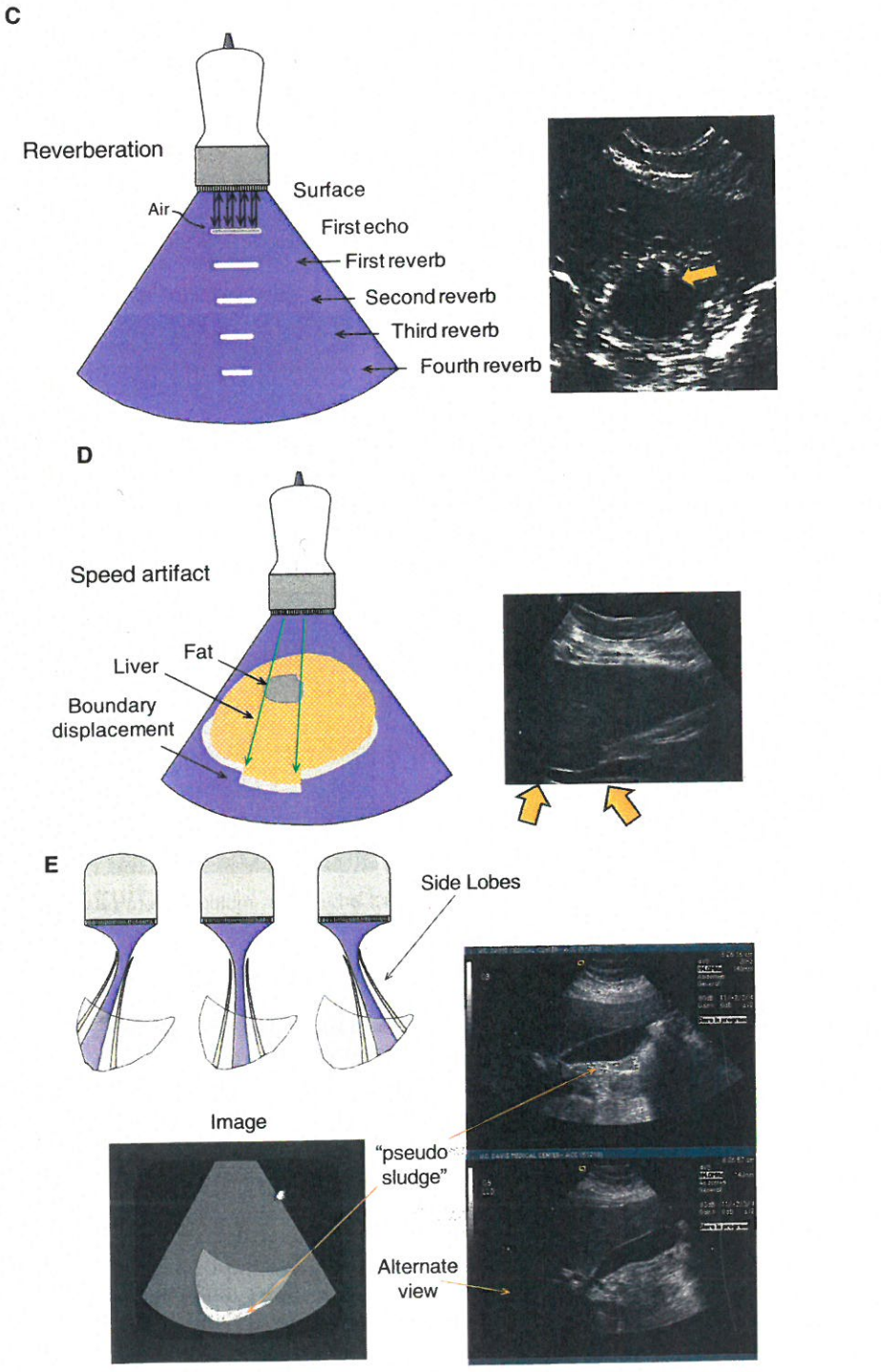
Reverberation

Reverberation artifacts arise from multiple echoes generated between two closely spaced interfaces reflecting ultrasound energy back and forth during the acquisition of the signal and before the next pulse. These artifacts are often caused by reflections between a highly reflective interface and the transducer or between reflective interfaces such as metallic objects (e.g., bullet fragments), calcified tissues, or air pocket/partial liquid areas of the anatomy. Reverberation echoes are typically manifested as multiple, equally spaced boundaries with decreasing amplitude along a straight line

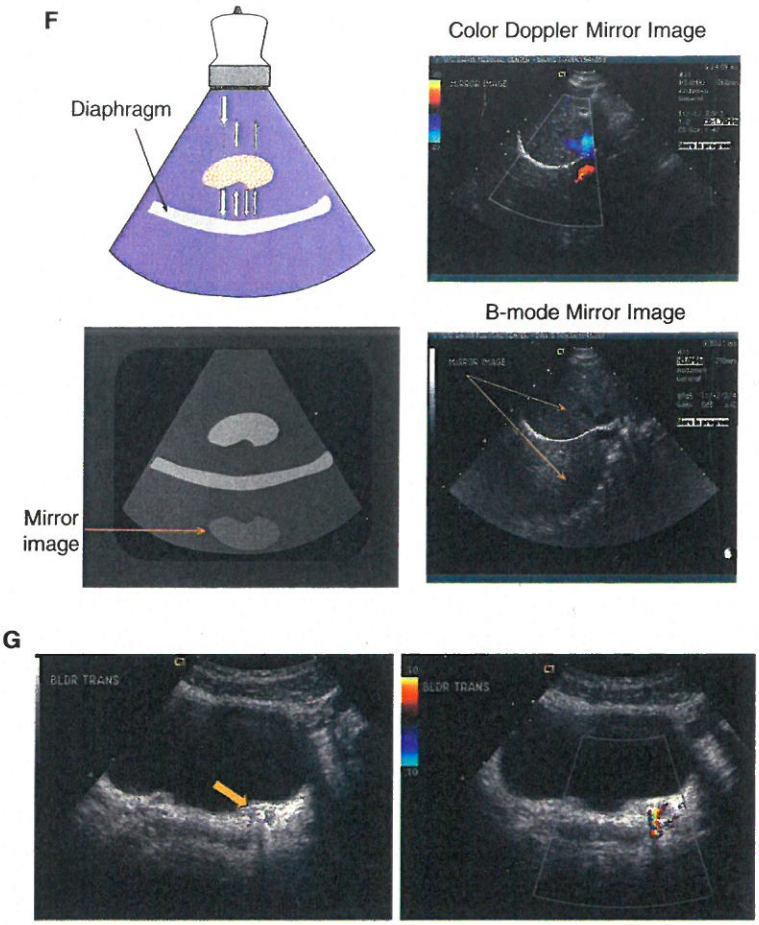
from the transducer (Fig. 14-52C). Comet tail artifact is a form of reverberation. Ring-down artifacts arise from resonant vibrations within fluid trapped between a tetrahedron of air bubbles, which creates a continuous sound wave that is transmitted back to the transducer and displayed as a series of parallel bands extending posterior to a collection of gas.



■ **FIGURE 14-52** **A.** Refraction is a change in the direction of the ultrasound beam that results from nonperpendicular incidence at a boundary where the two tissues do not have the same speed of sound. During a scan, anatomy can be missed and/or dislocated from the true position. **B.** Attenuation and reflection of the ultrasound beam cause intensity changes. Enhancement occurs distal to objects of low attenuation, manifested by a hyperintense signal, while shadowing occurs distal to objects of high attenuation, resulting in a hypointense signal. At the curved edges of an organ boundary or mass, nonperpendicular reflection can cause distal hypointense streaks and shadowing. Clinical images show through transmission (top) of a low attenuation cyst and shadowing (bottom), caused by high attenuation gallstones. **C.** Reverberation commonly occurs between to strong reflectors, such as an air pocket and the transducer array (left diagram) or with calcium deposits and internal reflections (right image). The echoes bounce back and forth between the two boundaries and produce equally spaced signals of diminishing amplitude in the image. This is often called a "comet-tail" artifact.



■ **FIGURE 14-52** (*Continued*) **D.** Speed of sound variation in the tissues can cause a mismapping of anatomy. In the case of fatty tissues, the slower speed of sound in fat (1,450 m/s) results in a displacement of the returning echoes from distal anatomy by about 6% of the distance traveled through the mass. **E.** Side lobe energy emissions in transducer arrays can cause anatomy outside of the main beam to be mapped into the main beam. For a curved boundary, such as the gallbladder, side lobe interactions can be remapped and produce findings such as "pseudo" sludge that is not apparent with other scanning angles. Clinical images (top) show pseudosludge, which is not evident after repositioning the transducer assembly (bottom).



■ **FIGURE 14-52** (*Continued*) **F.** A mirror image artifact arises from multiple beam reflections between a mass and a strong reflector, such as the diaphragm. Multiple echoes result in the creation of a mirror image of the mass beyond the diaphragm. Clinical images show mirror artifacts for color Doppler (**top**) and B-mode scan (**bottom**), where the lower arrow points to the artifact mirrored by the diaphragm. **G.** Twinkling artifact occurs with color flow imaging, caused by strong reflectors such as calcified stones, resulting in a changing mixture of colors indicating flow. On the left is a B-mode image of the bladder and the corresponding color flow image (***inset*** trapezoidal area) demonstrating the twinkling artifact.

Speed Displacement

The speed displacement artifact is caused by the variability of speed of sound in different tissues. In particular, the lower speed of sound in fat (1,450 *m/s*) causes edge discontinuities of organ borders distal to fatty tissue (Fig. 14-52D). Edges are mapped outward relative to the nondisplaced tissue borders by about 6% of the distance traveled through the fat (e.g., (1,540 – 1,450)/1,540 × 100 = 5.8%). Range and distance uncertainty result from the speed artifact, and this also reduces the accuracy of spatial measurements made with ultrasound.

Side Lobes and Grating Lobes

Side lobes are emissions of the ultrasound energy that occur in a direction slightly off-axis from the main beam (see Figs. 14-17 and 14-18) and arise from the expansion of the piezoelectric elements orthogonal to the main beam. Echoes returning from tissues along the propagation direction of the side lobes are positioned in the image

as if they occurred along the main beam. One type of side lobe artifact occurs near a highly reflective surface just out of the main beam. Sometimes, side lobes redirect diffuse echoes from adjacent soft tissues into an organ that is normally hypoechoic. This occurs, for instance, in imaging of the gall bladder, where the side lobes produce artifactual "pseudosludge" in an otherwise echo-free organ (Fig. 14-52E). Compared to grating lobes, side lobes are more forward directed and are present with all types of single-element and multielement transducer assemblies.

Grating lobes occur with multielement array transducers and result from the division of a smooth transducer surface into a large number of small elements. Ultrasound energy is produced at a large angle relative to the main beam. This misdirected energy can create ghost images of off-axis high-contrast objects. Grating lobe artifacts are reduced by using very closely spaced elements in the array (less than one-half wavelength apart). Linear array transducers are more prone to grating lobe artifacts than phased-array transducers, chiefly due to the larger width and spacing of the individual elements.

Multipath Reflection and Mirror Image

Near highly reflective surfaces, multiple beam reflections and refractions can find their way back to the transducer. The anatomy involved in these reflections is misplaced on the beam axis more distal to the actual position caused by delays of the echoes returning from the reflector(s). A common example is the interface of the liver and the diaphragm in abdominal imaging. The pulse from the transducer generates echoes from a mass in the liver and continues to the diaphragm, where a very strong echo is produced. This echo travels from the diaphragm back to the mass, producing another set of echoes now directed back to the diaphragm. These echoes are re-reflected from the diaphragm to the transducer. The back and forth travel distance of the second echo set from the mass produces an artifact in the image that resembles a mirror image of the mass, placed beyond the diaphragm (Fig. 14-52F).

Ambiguity

Ambiguity artifacts are created when a high PRF limits the amount of time spent listening for echoes during the PRP. As the PRF *increases*, the PRP *decreases*, with returning echoes still arriving from a greater depth after the next pulse is initiated. Mismapping of very deep echoes to shallow positions can occur in the image.

Twinkling Artifact

Doppler mode detects motion, particularly blood flow, and displays moving blood as red or blue (or selectable color) on the monitor, depending on direction and velocity. The twinkling artifact is represented as a rapidly changing mixture of colors, is typically seen distal to a strong reflector such as a calculus, and is often mistaken for an aneurysm when evaluating vessels. This artifactual appearance is possibly due to echoes from the strong reflector with frequency changes due to the wide bandwidth of the initial pulse and the narrow band "ringing" caused by the structure. Twinkling artifact may be used to identify small renal stones, as shown in Figure 14-52G, and differentiate echogenic foci from calcifications within kidney, gall bladder, and liver.

Slice Thickness

Slice thickness is determined by the beam width of the transducer array perpendicular to the image plane and is greater than the beam width in the image plane. The thickness of the slice profile varies with depth, being broad close to the transducer, narrowing at the elevational focal zone, and widening with distance beyond the focal