CHAPTER

19

ULTRASOUND WAVES

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OBJECTIVES

From studying this chapter, the reader should be able to:

- Explain the properties of ultrasound waves.
- Describe the decibel notation for ultrasound intensity and pressure.
- Delineate the ultrasound properties of velocity, attenuation, and absorption.
- Depict the consequences of an impedance mismatch at the boundary between two regions of tissue.
- Explain ultrasound reflection, refraction and scattering.

INTRODUCTION

Ultrasound is a mechanical disturbance that moves as a pressure wave through a medium. When the medium is a patient, the wavelike disturbance is the basis for use of ultrasound as a diagnostic tool. Appreciation of the characteristics of ultrasound waves and their behavior in various media is essential to understanding the use of diagnostic ultrasound in clinical medicine.[1–6]

In 1794, Spallanzi suggested correctly that bats avoided obstacles during flight by using sound signals beyond the range of the human ear.

HISTORY

In 1880, French physicists Pierre and Jacques Curie discovered the piezoelectric effect.[7] French physicist Paul Langevin attempted to develop piezoelectric materials as senders and receivers of high-frequency mechanical disturbances (ultrasound waves) through materials.[8] His specific application was the use of ultrasound to detect submarines during World War I. This technique, sound navigation and ranging (SONAR), finally became practical during World War II. Industrial uses of ultrasound began in 1928 with the suggestion of Soviet Physicist Sokolov that it could be used to detect hidden flaws in materials. Medical uses of ultrasound through the 1930s were confined to therapeutic applications such as cancer treatments and physical therapy for various ailments. Diagnostic applications of ultrasound began in the late 1940s through collaboration between physicians and engineers familiar with SONAR.[9]

"Piezo" is Greek for pressure. *Piezoelectricity* refers to the generation of an electrical response to applied pressure.

Pierre Curie used the piezoelectric properties of quartz crystals to construct a device to measure the small changes in mass that accompany radioactive decay. This work was done in collaboration with his wife Marie in her early studies of radioactivity.

The term "transducer" refers to any device that converts energy from one form to another (mechanical to electrical, electrical to heat, etc.). When someone asks to see a "transducer" in a radiology department, they will be shown ultrasound equipment. But strictly speaking, they could just as well be taken to see an x-ray tube.

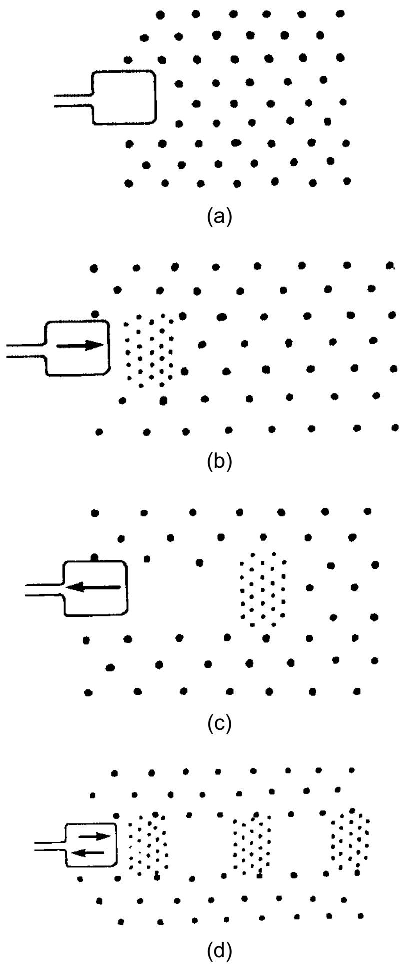
WAVE MOTION

A fluid medium is a collection of molecules that are in continuous random motion. The molecules are represented as filled circles in the margin figure (Margin Figure 19-1). When no external force is applied to the medium, the molecules are distributed more or less uniformly (**A**). When a force is applied to the medium (represented by movement of the piston from left to right in **B**), the molecules are concentrated in front of the piston, resulting in an increased pressure at that location. The region of increased pressure is termed a *zone of compression*. Because of the forward motion imparted to the molecules by the piston, the region of increased pressure begins to migrate away from the piston and through the medium. That is, a mechanical disturbance introduced into the medium travels through the medium in a direction away from the source of the disturbance. In clinical applications of ultrasound, the piston is replaced by an ultrasound transducer.

As the zone of compression begins its migration through the medium, the piston may be withdrawn from right to left to create a region of reduced pressure immediately behind the compression zone. Molecules from the surrounding medium move into this region to restore it to normal particle density; and a second region, termed a *zone of rarefaction*, begins to migrate away from the piston (**C**). That is, the compression zone (high pressure) is followed by a zone of rarefaction (low pressure) also moving through the medium.

If the piston is displaced again to the right, a second compression zone is established that follows the zone of rarefaction through the medium, If the piston oscillates continuously, alternate zones of compression and rarefaction are propagated through the medium, as illustrated in **D**. The propagation of these zones establishes a wave disturbance in the medium. This disturbance is termed a *longitudinal wave* because the motion of the molecules in the medium is parallel to the direction of wave propagation. A wave with a frequency between about 20 and 20,000 Hz is a sound wave that is audible to the human ear. An infrasonic wave is a sound wave below 20 Hz; it is not audible to the human ear. An ultrasound (or ultrasonic) wave has a frequency greater than 20,000 Hz and is also inaudible. In clinical diagnosis, ultrasound waves of frequencies between 1 and 20 MHz are used.

As a longitudinal wave moves through a medium, molecules at the edge of the wave slide past one another. Resistance to this shearing effect causes these molecules to move somewhat in a direction away from the moving longitudinal wave. This transverse motion of molecules along the edge of the longitudinal wave establishes shear waves that radiate transversely from the longitudinal wave. In general, shear waves are significant only in a rigid medium such as a solid. In biologic tissues, bone is the only medium in which shear waves are important.



MARGIN FIGURE 19-1
Production of an ultrasound wave. **A:** Uniform distribution of molecules in a medium. **B:** Movement of the piston to the right produces a zone of compression. **C:** Withdrawal of the piston to the left produces a zone of rarefaction. **D:** Alternate movement of the piston to the right and left establishes a longitudinal wave in the medium.

WAVE CHARACTERISTICS

A zone of compression and an adjacent zone of rarefaction constitute one cycle of an ultrasound wave. A wave cycle can be represented as a graph of local pressure (particle density) in the medium versus distance in the direction of the ultrasound wave (Figure 19-1). The distance covered by one cycle is the wavelength of the ultrasound wave. The number of cycles per unit time (cps, or just sec^{-1}) introduced into the medium each second is referred to as the *frequency of the wave*, expressed in units of hertz, kilohertz, or megahertz where 1 Hz equals 1 cps. The maximum height of the wave cycle is the amplitude of the ultrasound wave. The product of the frequency (ν) and the wavelength (λ) is the velocity of the wave; that is, $c = \nu\lambda$.

In most soft tissues, the velocity of ultrasound is about 1540 m/sec. Frequencies of 1 MHz and greater are required to furnish ultrasound wavelengths suitable for diagnostic imaging.

When two waves meet, they are said to "interfere" with each other (see Margin). There are two extremes of interference. In constructive interference the waves are "in phase" (i.e., peak meets peak). In destructive interference the waves are "out of phase" (i.e., peak meets valley). Waves undergoing constructive interference add their amplitudes, whereas waves undergoing destructive interference may completely cancel each other.

During the propagation of an ultrasound wave, the molecules of the medium vibrate over very short distances in a direction parallel to the longitudinal wave. It is this vibration, during which momentum is transferred among molecules, that causes the wave to move through the medium.

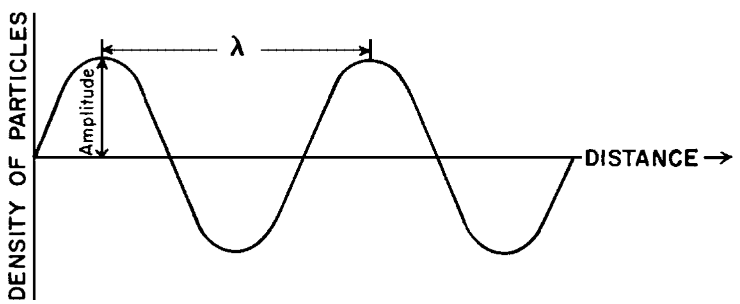
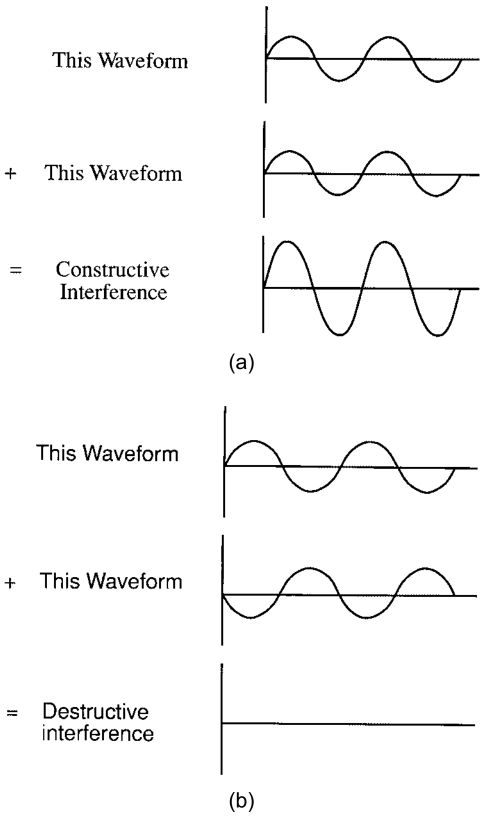


FIGURE 19-1
Characteristics of an ultrasound wave.

Frequency Classification of Ultrasound

Frequency (Hz)	*Classification*
20–20,000	Audible sound
20,000–1,000,000	Ultrasound
1,000,000–30,000,000	Diagnostic medical ultrasound

Ultrasound frequencies of 1 MHz and greater correspond to ultrasound wavelengths less than 1 mm in human soft tissue.



MARGIN FIGURE 19-2
Waves can exhibit interference, which in extreme cases of constructive and destructive interference leads to complete addition (**A**) or complete cancellation (**B**) of the two waves.

In the audible range, sound power and intensity are referred to as "loudness."

Pulsed ultrasound is used for most medical diagnostic applications. Ultrasound pulses vary in intensity and time and are characterized by four variables: spatial peak (SP), spatial average (SA), temporal peak (TP), and temporal average (TA).

Temporal average ultrasound intensities used in medical diagnosis are in the mW/cm^2 range.

TABLE 19-1 Quantities and Units Pertaining to Ultrasound Intensity

Quantity	*Definition*	*Unit*
Energy (E)	Ability to do work	joule
Power (P)	Rate at which energy is transported	watt (joule/sec)
Intensity (I)	Power per unit area (a), where t = time	$watt/cm^2$
Relationship	$I = \frac{P}{a} = \frac{E}{(t)(a)}$	

ULTRASOUND INTENSITY

As an ultrasound wave passes through a medium, it transports energy through the medium. The rate of energy transport is known as "power." Medical ultrasound is produced in beams that are usually focused into a small area, and the beam is described in terms of the power per unit area, defined as the beam's "intensity." The relationships among the quantities and units pertaining to intensity are summarized in Table 19-1.

Intensity is usually described relative to some reference intensity. For example, the intensity of ultrasound waves sent into the body may be compared with that of the ultrasound reflected back to the surface by structures in the body. For many clinical situations the reflected waves at the surface may be as much as a hundredth or so of the intensity of the transmitted waves. Waves reflected from structures at depths of 10 cm or more below the surface may be lowered in intensity by a much larger factor. A logarithmic scale is most appropriate for recording data over a range of many orders of magnitude. In acoustics, the decibel scale is used, with the decibel defined as

$$\text{dB} = 10 \log \frac{I}{I_0} \qquad \textbf{(19-1)}$$

where I_0 is the reference intensity. Table 19-2 shows examples of decibel values for certain intensity ratios. Several rules can be extracted from this table:

- Positive decibel values result when a wave has a higher intensity than the reference wave; negative values denote a wave with lower intensity.
- Increasing a wave's intensity by a factor of 10 adds 10 dB to the intensity, and reducing the intensity by a factor of 10 subtracts 10 dB.
- Doubling the intensity adds 3 dB, and halving subtracts 3 dB.

No universal standard reference intensity exists for ultrasound. Thus the statement "ultrasound at 50 dB was used" is nonsensical. However, a statement such as "the returning echo was 50 dB below the transmitted signal" is informative. The transmitted signal then becomes the reference intensity for this particular application. For

TABLE 19-2 Calculation of Decibel Values From Intensity Ratios and Amplitude Ratios

Ratio of Ultrasound Wave Parameters	*Intensity Ratio (I/I_0) (dB)*	*Amplitude Ratio (A/A_0) (dB)*
1000	30	60
100	20	40
10	10	20
2	3	6
1	0	0
1/2	−3	−6
1/10	−10	−20
1/100	−20	−40
1/1000	−30	−60

audible sound, a statement such as "a jet engine produces sound at 100 dB" is appropriate because there is a generally accepted reference intensity of 10^{-16} W/cm^2 for audible sound.[10] A 1-kHz tone (musical note C one octave above middle C) at this intensity is barely audible to most listeners. A 1-kHz note at 120 dB (10^{-4} W/cm^2) is painfully loud.

The human ear is unable to distinguish a difference in loudness less than about 1 dB.

Because intensity is power per unit area and power is energy per unit time (Table 19-1), Eq. (19-1) may be used to compare the power or the energy contained within two ultrasound waves. Thus we could also write

$$\text{dB} = 10 \log \frac{\text{Power}}{\text{Power}_0} = 10 \log \frac{E}{E_0}$$

Ultrasound wave intensity is related to maximum pressure (P_m) in the medium by the following expression[11]:

$$1 = \frac{P_m^2}{2\rho c} \qquad \textbf{(19-2)}$$

where ρ is the density of the medium in grams per cubic centimeter and c is the speed of sound in the medium. Substituting Eq. (19-2) for I and I_0 in Eq. (19-1) yields

$$\text{dB} = 10 \log \frac{P_m^2/2\rho c}{(P_m^2)_0/2\rho c} = 10 \log \left[\frac{P_m}{P_{m_0}}\right]^2$$

$$= 20 \log \frac{P_m}{P_{m_0}} \qquad \textbf{(19-3)}$$

Calculation of Decibel Value from Wave Parameters

For X	= Intensity in W/cm^2	
	= Power in watts	
	= Energy in joules	
use dB	$= 10 \log \frac{x}{x_0}$	
For Y	= Pressure in pascals or atmospheres	
	= Amplitude in volts	
Use dB	$= 20 \log \frac{Y}{Y_o}$	

When comparing the pressure of two waves, Eq. (19-3) may be used directly. That is, the pressure does not have to be converted to intensity to determine the decibel value. An ultrasound transducer converts pressure amplitudes received from the patient (i.e., the reflected ultrasound wave) into voltages. The amplitude of voltages recorded for ultrasound waves is directly proportional to the variations in pressure in the reflected wave.

The decibel value for the ratio of two waves may be calculated from Eq. (19-1) or from Eq. (19-3), depending upon the information that is available concerning the waves (see Margin Table). The "half-power value" (ratio of 0.5 in power between two waves) is –3 dB, whereas the "half-amplitude value" (ratio of 0.5 in amplitude) is –6 dB (Table 19-2). This difference reflects the greater sensitivity of the decibel scale to amplitude compared with intensity values.

Ultrasound intensities may also be compared in units of nepers per centimeter by using the natural logarithm (ln) rather than the common logarithm (log) where

$$\text{Neper} = \ln (I/I_0)$$

ULTRASOUND VELOCITY

The velocity of an ultrasound wave through a medium varies with the physical properties of the medium. In low-density media such as air and other gases, molecules may move over relatively large distances before they influence neighboring molecules. In these media, the velocity of an ultrasound wave is relatively low. In solids, molecules are constrained in their motion, and the velocity of ultrasound is relatively high. Liquids exhibit ultrasound velocities intermediate between those in gases and solids. With the notable exceptions of lung and bone, biologic tissues yield velocities roughly similar to the velocity of ultrasound in liquids. In different media, changes in velocity are reflected in changes in wavelength of the ultrasound waves, with the frequency remaining relatively constant. In ultrasound imaging, variations in the velocity of ultrasound in different media introduce artifacts into the image, with the major artifacts attributable to bone, fat, and, in ophthalmologic applications, the lens of the eye. The velocities of ultrasound in various media are listed in Table 19-3.

In ultrasound, the term *propogation speed* is preferred over the term *velocity*.

The velocity of an ultrasound wave should be distinguished from the velocity of molecules whose displacement into zones of compression and rarefaction constitutes the wave. The molecular velocity describes the velocity of the individual molecules in the medium, whereas the wave velocity describes the velocity of the ultrasound wave

The velocity of ultrasound in a medium is virtually independent of the ultrasound frequency.

TABLE 19-3 Approximate Velocities of Ultrasound in Selected Materials

Nonbiologic Material	*Velocity (m/sec)*	*Biologic Material*	*Velocity (m/sec)*
Acetone	1174	Fat	1475
Air	331	Brain	1560
Aluminum (rolled)	6420	Liver	1570
Brass	4700	Kidney	1560
Ethanol	1207	Spleen	1570
Glass (Pyrex)	5640	Blood	1570
Acrylic plastic	2680	Muscle	1580
Mercury	1450	Lens of eye	1620
Nylon (6-6)	2620	Skull bone	3360
Polyethylene	1950	Soft tissue (mean value)	1540
Water (distilled), 25°C	1498		
Water (distilled), 50°C	1540		

The velocity of ultrasound is determined principally by the compressibility of the medium. A medium with high compressibility yields a slow ultrasound velocity, and vice versa. Hence, the velocity is relatively low in gases, intermediate in soft tissues, and greatest in solids such as bone.

through the medium. Properties of ultrasound such as reflection, transmission, and refraction are characteristic of the wave velocity rather than the molecular velocity.

ATTENUATION OF ULTRASOUND

As an ultrasound beam penetrates a medium, energy is removed from the beam by absorption, scattering, and reflection. These processes are summarized in Figure 19-2. As with x rays, the term *attenuation* refers to any mechanism that removes energy from the ultrasound beam. Ultrasound is "absorbed" by the medium if part of the beam's

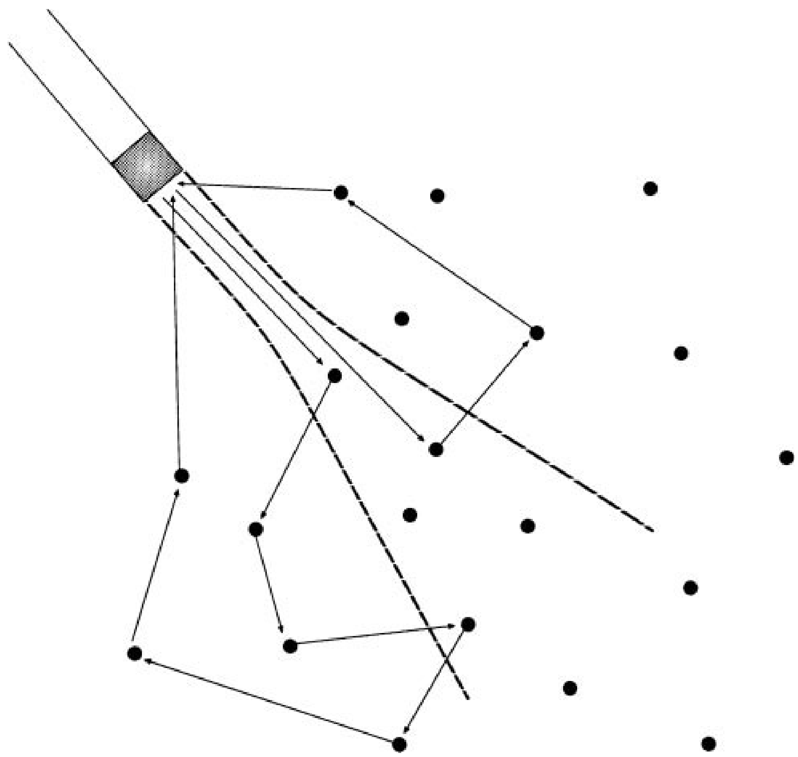


FIGURE 19-2
Constructive and destructive interference effects characterize the echoes from nonspecular reflections. Because the sound is reflected in all directions, there are many opportunities for waves to travel different pathways. The wave fronts that return to the transducer may constructively or destructively interfere at random. The random interference pattern is known as "speckle."

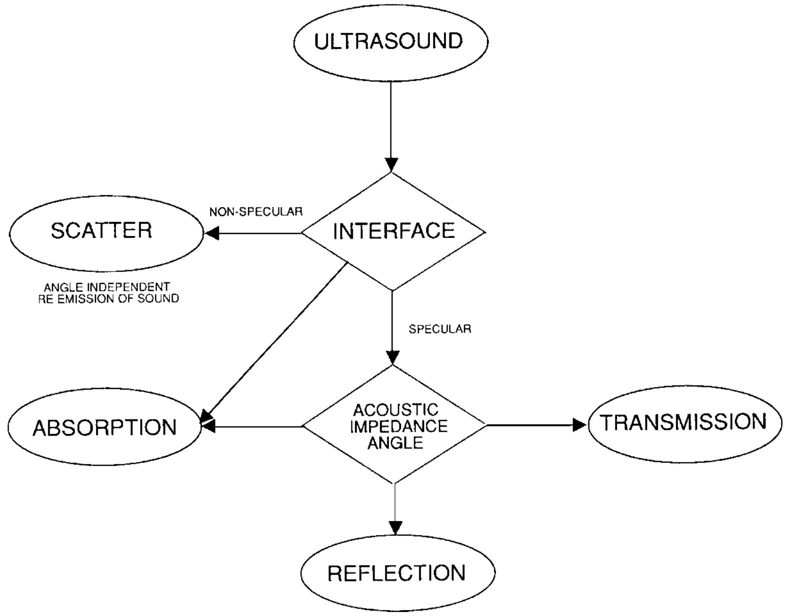


FIGURE 19-3
Summary of interactions of ultrasound at boundaries of materials.

energy is converted into other forms of energy, such as an increase in the random motion of molecules. Ultrasound is "reflected" if there is an orderly deflection of all or part of the beam. If part of an ultrasound beam changes direction in a less orderly fashion, the event is usually described as "scatter."

The behavior of a sound beam when it encounters an obstacle depends upon the size of the obstacle compared with the wavelength of the sound. If the obstacle's size is large compared with the wavelength of sound (and if the obstacle is relatively smooth), then the beam retains its integrity as it changes direction. Part of the sound beam may be reflected and the remainder transmitted through the obstacle as a beam of lower intensity.

If the size of the obstacle is comparable to or smaller than the wavelength of the ultrasound, the obstacle will scatter energy in various directions. Some of the ultrasound energy may return to its original source after "nonspecular" scatter, but probably not until many scatter events have occurred.

In ultrasound imaging, specular reflection permits visualization of the boundaries between organs, and nonspecular reflection permits visualization of tissue parenchyma (Figure 19-2). Structures in tissue such as collagen fibers are smaller than the wavelength of ultrasound. Such small structures provide scatter that returns to the transducer through multiple pathways. The sound that returns to the transducer from such nonspecular reflectors is no longer a coherent beam. It is instead the sum of a number of component waves that produces a complex pattern of constructive and destructive interference back at the source. This interference pattern, known as "speckle," provides the characteristic ultrasonic appearance of complex tissue such as liver.

The behavior of a sound beam as it encounters an obstacle such as an interface between structures in the medium is summarized in Figure 19-3. As illustrated in Figure 19-4, the energy remaining in the beam decreases approximately exponentially with the depth of penetration of the beam into the medium. The reduction in energy (i.e., the decrease in ultrasound intensity) is described in decibels, as noted earlier.

Example 19-1

Find the percent reduction in intensity for a 1-MHz ultrasound beam traversing 10 cm of material having an attenuation of 1 dB/cm. The reduction in intensity (dB) = (1 dB/cm) (10 cm) = −10 dB (the minus sign corresponds to a decrease in intensity

An increase in the random motion of molecules is measurable as an increase in temperature of the medium.

Contributions to attenuation of an ultrasound beam may include:

- Absorption
- Reflection
- Scattering
- Refraction
- Diffraction
- Interference
- Divergence

Reflection in which the ultrasound beam retains its integrity is said to be "specular," from the Latin for "mirror." Reflection of visible light from a plane mirror is an optical example of specular reflection (i.e., the original shape of the wave fronts).

An optical example of nonspecular reflection occurs when a mirror is "steamed up." The water droplets on the mirror are nonspecular reflectors that serve to scatter the light beam.

When scattering objects are much smaller than the ultrasound wavelength, the scattering process is referred to as Rayleigh scattering. Red blood cells are sometimes referred to as Rayleigh scatterers.

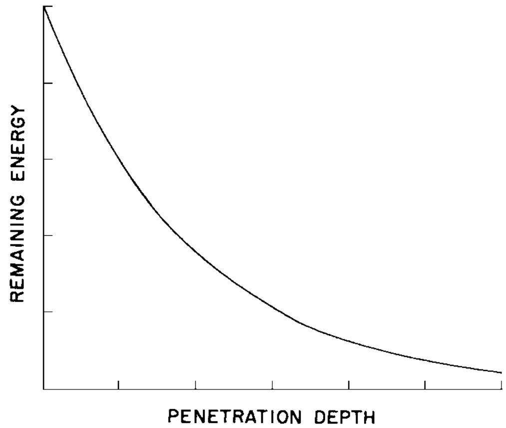


FIGURE 19-4
Energy remaining in an ultrasound beam as a function of the depth of penetration of the beam into a medium.

compared with the reference intensity, which in this case is the intensity of sound before the attenuating material is encountered).

$$dB = 10 \log \frac{I}{I}$$

$$-10 = 10 \log \frac{I}{I_0}$$

$$1 = \log \frac{I_0}{I}$$

$$10 = \frac{I_0}{I}$$

$$I = \frac{I_0}{10}$$

There has been a 90% reduction in intensity. Determine the intensity reduction if the ultrasound frequency were increased to 2MHz.

Because the attenuation increases approximately linearly with frequency, the attenuation coefficient at 2 MHz would be 2 dB/cm, resulting in a –20 dB (99%) intensity reduction.

The attenuation of ultrasound in a material is described by the attenuation coefficient α in units of decibels per centimeter (Table 19-4). Many of the values in Table 19-4 are known only approximately and vary significantly with both the origin and condition of the biologic samples. The attenuation coefficient α is the sum of the individual coefficients for scatter and absorption. In soft tissue, the absorption

TABLE 19-4 Attenuation Coefficients α for 1-MHz Ultrasound

Material	*α (dB/cm)*	*Material*	*α (dB/cm)*
Blood	0.18	Lung	40
Fat	0.6	Liver	0.9
Muscle (across fibers)	3.3	Brain	0.85
Muscle (along fibers)	1.2	Kidney	1.0
Aqueous and vitreous humor of eye	0.1	Spinal cord	1.0
		Water	0.0022
Lens of eye	2.0	Caster oil	0.95
Skull bone	20	Lucite	2.0

coefficient accounts for 60% to 90% of the attenuation, and scatter accounts for the remainder.[11]

Table 19-4 shows that the attenuation of ultrasound is very high in bone. This property, along with the large reflection coefficient of a tissue-bone interface, makes it difficult to visualize structures lying behind bone. Little attenuation occurs in water, and this medium is a very good transmitter of ultrasound energy. To a first approximation, the attenuation coefficient of most soft tissues can be approximated as 0.9υ, where υ is the frequency of the ultrasound in MHz. This expression states that the attenuation of ultrasound energy increases with frequency in biologic tissues. That is, higher-frequency ultrasound is attenuated more readily and is less penetrating than ultrasound of lower frequency.

The energy loss in a medium composed of layers of different materials is the sum of the energy loss in each layer.

Attenuation causes a loss of signal intensity when structures at greater depths are imaged with ultrasound. This loss of intensity is compensated by increasing the amplification of the signals.

Differences in attenuation among tissues causes enhancement and shadowing in ultrasound images.

Example 19-2

Suppose that a block of tissue consists of 2 cm fat, 3 cm muscle (ultrasound propagated parallel to the fibers), and 4 cm liver. The total energy loss is

$$\begin{aligned}\text{Total energy loss} &= (\text{Energy loss in fat}) + (\text{Energy loss in muscle})\\ &\quad + (\text{Energy loss in liver})\\ &= (0.6\ \text{dB/cm})\ (2\ \text{cm}) + (1.2\ \text{dB/cm})\ (3\ \text{cm})\\ &\quad + (09\ \text{dB/cm})\ (4\ \text{cm})\\ &= 1.2\ \text{dB} + 3.6\ \text{dB} + 3.6\ \text{dB}\\ &= 8.4\ \text{dB}\end{aligned}$$

For an ultrasound beam that traverses the block of tissue and, after reflection, returns through the tissue block, the total attenuation is twice 8.4 dB or 16.8 dB.

REFLECTION

In most diagnostic applications of ultrasound, use is made of ultrasound waves reflected from interfaces between different tissues in the patient. The fraction of the impinging energy reflected from an interface depends on the difference in acoustic impedance of the media on opposite sides of the interface.

The acoustic impedance Z of a medium is the product of the density ρ of the medium and the velocity of ultrasound in the medium:

$$Z = \rho c$$

Acoustic impedances of several materials are listed in the margin. For an ultrasound wave incident perpendicularly upon an interface, the fraction α_R of the incident energy that is reflected (i.e., the reflection coefficient α_R) is

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

where Z_1 and Z_2 are the acoustic impedances of the two media. The fraction of the incident energy that is transmitted across an interface is described by the transmission coefficient α_T, where

$$\alpha_T = \frac{4Z_1Z_2}{(Z_1 + Z_2)^2}$$

Obviously $\alpha_T + \alpha_R = 1$.

In this discussion, reflection is assumed to occur at interfaces that have dimensions greater than the ultrasound wavelength. In this case, the reflection is termed *specular reflection.*

Acoustic impedance may be expressed in units of rayls, where a rayl = 1 kg-m^{-2}-sec^{-1}.

The rayl is named for Lord Rayleigh [John Strutt (1842–1919), the 3rd Baron Rayleigh], a British physicist who pioneered the study of molecular motion in gasses that explains sound propogation.

Approximate Acoustic Impedances of Selected Materials

Material	*Acoustic Impedance* *(kg-m^{-2} -sec^{-1})* $\times 10^{-4}$
Air at standard temperature and pressure	0.0004
Water	1.50
Polyethylene	1.85
Plexiglas	3.20
Aluminum	18.0
Mercury	19.5
Brass	38.0
Fat	1.38
Aqueous and vitreous humor of eye	1.50
Brain	1.55
Blood	1.61
Kidney	1.62
Human soft tissue, mean value	1.63
Spleen	1.64
Liver	1.65
Muscle	1.70
Lens of eye	1.85
Skull bone	6.10

With a large impedance mismatch at an interface, much of the energy of an ultrasound wave is reflected, and only a small amount is transmitted across the interface. For example, ultrasound beams are reflected strongly at air–tissue and air–water interfaces because the impedance of air is much less than that of tissue or water.

Example 19-3

At a "liver–air" interface, $Z_1 = 1.65$ and $Z_2 = 0.0004$ (both multiplied by 10^{-4} with units of kg-m^{-2}-sec^{-1}).

$$\alpha_R = \left(\frac{1.65 - 0.0004}{1.65 + 0.0004}\right)^2, \qquad \alpha_T = \frac{4(1.65)(0.0004)}{(1.65 + 0.0004)^2}$$
$$= 0.9995 \qquad\qquad = 0.0005$$

Thus 99.95% of the ultrasound energy is reflected at the air–liver interface, and only 0.05% of the energy is transmitted. At a muscle ($Z = 1.70$)–liver ($Z = 1.65$) interface,

$$\alpha_R = \left(\frac{1.70 - 1.65}{1.70 + 1.65}\right)^2, \qquad \alpha_T = \frac{4(1.70)(1.65)}{(1.70 + 1.65)^2}$$
$$= 0.015 \qquad\qquad = 0.985$$

At a muscle–liver interface, slightly more than 1% of the incident energy is reflected, and about 99% of the energy is transmitted across the interface. Even though the reflected energy is small, it is often sufficient to reveal the liver border. The magnitudes of echoes from various interfaces in the body are described in Figure 19-5.

Because of the high value of the coefficient of ultrasound reflection at an air–tissue interface, water paths and various creams and gels are used during ultrasound examinations to remove air pockets (i.e., to obtain good acoustic coupling) between the ultrasound transducer and the patient's skin. With adequate acoustic coupling,

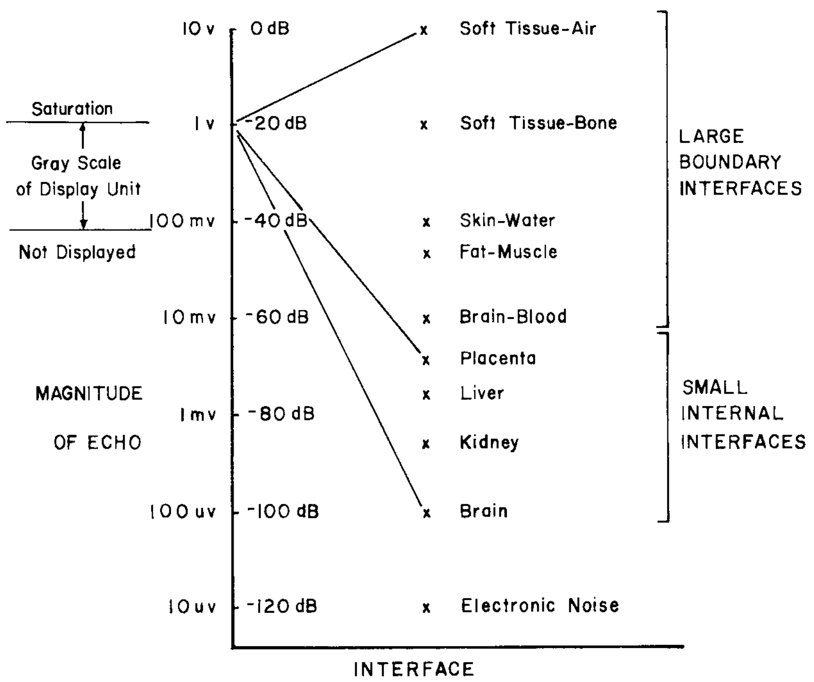
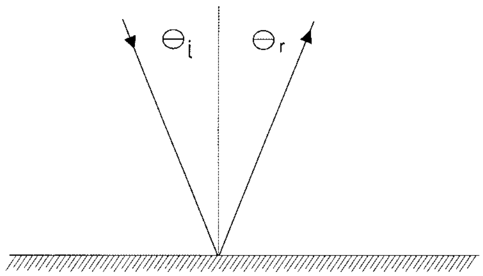


FIGURE 19-5
Range of echoes from biologic interfaces and selection of internal echoes to be displayed over the major portion of the gray scale in an ultrasound unit. (From Kossoff, G., et al.[12] Used with permission).

the ultrasound waves will enter the patient with little reflection at the skin surface. Similarly, strong reflections of ultrasound occur at the boundary between the chest wall and the lungs and at the millions of air–tissue interfaces within the lungs. Because of the large impedance mismatch at these interfaces, efforts to use ultrasound as a diagnostic tool for the lungs have been unrewarding. The impedance mismatch is also high between soft tissues and bone, and the use of ultrasound to identify tissue characteristics in regions behind bone has had limited success.

The discussion of ultrasound reflection above assumes that the ultrasound beam strikes the reflecting interface at a right angle. In the body, ultrasound impinges upon interfaces at all angles. For any angle of incidence, the angle at which the reflected ultrasound energy leaves the interface equals the angle of incidence of the ultrasound beam; that is,

$$\text{Angle of incidence} = \text{Angle of reflection}$$



MARGIN FIGURE 19-3
Ultrasound reflection at an interface, where the angle of incidence θ_i equals the angle of reflection θ_r.

In a typical medical examination that uses reflected ultrasound and a transducer that both transmits and detects ultrasound, very little reflected energy will be detected if the ultrasound strikes the interface at an angle more than about 3 degrees from perpendicular. A smooth reflecting interface must be essentially perpendicular to the ultrasound beam to permit visualization of the interface.

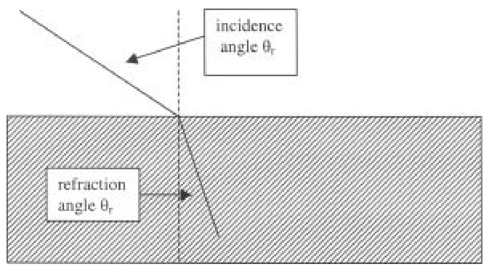
Two conditions are required for refraction to occur: (1) The sound beam must strike an interface at an angle other than 90°; (2) the speed of sound must differ on opposite sides of the interface.

REFRACTION

As an ultrasound beam crosses an interface obliquely between two media, its direction is changed (i.e., the beam is bent). If the velocity of ultrasound is higher in the second medium, then the beam enters this medium at a more oblique (less steep) angle. This behavior of ultrasound transmitted obliquely across an interface is termed *refraction*. The relationship between incident and refraction angles is described by Snell's law:

$$\frac{\text{Sine of incidence angle}}{\text{Sine of refractive angle}} = \frac{\begin{array}{c}\text{Velocity in}\\ \text{incidence medium}\end{array}}{\begin{array}{c}\text{Velocity in}\\ \text{refractive medium}\end{array}} \tag{19-4}$$

$$\frac{\sin\theta_i}{\sin\theta_r} = \frac{c_i}{c_r}$$



MARGIN FIGURE 19-4
Refraction of ultrasound at an interface, where the ratio of the velocities of ultrasound in the two media is related to the sine of the angles of incidence and refraction.

For example, an ultrasound beam incident obliquely upon an interface between muscle (velocity 1580 m/sec) and fat (velocity 1475 m/sec) will enter the fat at a steeper angle.

If an ultrasound beam impinges very obliquely upon a medium in which the ultrasound velocity is higher, the beam may be refracted so that no ultrasound energy enters the medium. The incidence angle at which refraction causes no ultrasound to enter a medium is termed the *critical angle* θ_c. For the critical angle, the angle of refraction is 90 degrees, and the sine of 90 degrees is 1. From Eq. (19-4),

$$\frac{\sin\theta_c}{\sin 90^\circ} = \frac{c_i}{c_r}$$

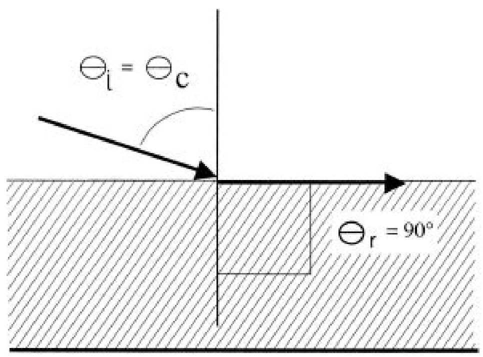
but

$$\sin 90^\circ = 1$$

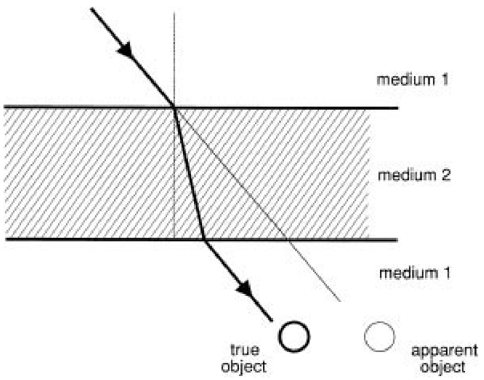
therefore

$$\theta_c = \sin^{-1}[c_i/c_r]$$

where $\sin^{-1}$, or arcsin, refers to the angle whose sine is c_i/c_r. For any particular interface, the critical angle depends only upon the velocity of ultrasound in the two media separated by the interface.



MARGIN FIGURE 19-5
For an incidence angle θ_c equal to the critical angle, refraction causes the sound to be transmitted along the surface of the material. For incidence angles greater than θ_c, sound transmission across the interface is prevented by refraction.



MARGIN FIGURE 19-6
Lateral displacement of an ultrasound beam as it traverses a slab interposed in an otherwise homogeneous medium.

Refraction is a principal cause of artifacts in clinical ultrasound images. In Margin Figure 19-6, for example, the ultrasound beam is refracted at a steeper angle as it crosses the interface between medium 1 and 2 ($c_1 > c_2$). As the beam emerges from medium 2 and reenters medium 1, it resumes its original direction of motion. The presence of medium 2 simply displaces the ultrasound beam laterally for a distance that depends upon the difference in ultrasound velocity and density in the two media and upon the thickness of medium 2. Suppose a small structure below medium 2 is visualized by reflected ultrasound. The position of the structure would appear to the viewer as an extension of the original direction of the ultrasound through medium 1. In this manner, refraction adds spatial distortion and resolution loss to ultrasound images.

Maximum ultrasound intensities (mW/cm^2) recommended by the U.S. Food and Drug Administration for various diagnostic applications. The values are spatial-peak, temporal-average (SPTA) values.[5,13]

Use	$(Intensity)_{max}$
Cardiac	430
Peripheral vessels	720
Ophthalmic	17
Abdominal	94
Fetal	94

ABSORPTION

Relaxation processes are the primary mechanisms of energy dissipation for an ultrasound beam transversing tissue. These processes involve (a) removal of energy from the ultrasound beam and (b) eventual dissipation of this energy primarily as heat. As discussed earlier, ultrasound is propagated by displacement of molecules of a medium into regions of compression and rarefaction. This displacement requires energy that is provided to the medium by the source of ultrasound. As the molecules attain maximum displacement from an equilibrium position, their motion stops, and their energy is transformed from kinetic energy associated with motion to potential energy associated with position in the compression zone. From this position, the molecules begin to move in the opposite direction, and potential energy is gradually transformed into kinetic energy. The maximum kinetic energy (i.e., the highest molecular velocity) is achieved when the molecules pass through their original equilibrium position, where the displacement and potential energy are zero. If the kinetic energy of the molecule at this position equals the energy absorbed originally from the ultrasound beam, then no dissipation of energy has occurred, and the medium is an ideal transmitter of ultrasound. Actually, the conversion of kinetic to potential energy (and vice versa) is always accompanied by some dissipation of energy. Therefore, the energy of the ultrasound beam is gradually reduced as it passes through the medium. This reduction is termed *relaxation energy loss*. The rate at which the beam energy decreases is a reflection of the attenuation properties of the medium.

The effect of frequency on the attenuation of ultrasound in different media is described in Table 19-5.[14–18] Data in this table are reasonably good estimates of the influence of frequency on ultrasound absorption over the range of ultrasound frequencies used diagnostically. However, complicated structures such as tissue samples often exhibit a rather complex attenuation pattern for different frequencies, which probably reflects the existence of a variety of relaxation frequencies and other molecular energy absorption processes that are poorly understood at present. These complex attenuation patterns are reflected in the data in Figure 19-6.

American Institute of Ultrasound in Medicine Safety Statement on Diagnostic Ultrasound has been in use since the early 1950s. Given its known benefits and recognized efficacy for medical diagnosis, including use during human pregnancy, the American Institute of Ultrasound in Medicine hereby addresses the clinical safety of such use as follows:

> No confirmed biological effects on patients or instrument operators caused by exposure at intensities typical of present diagnostic ultrasound instruments have ever been reported. Although the possibility exists that such biological effects may be identified in the future, current data indicate that the benefits to patients of the prudent use of diagnostic ultrasound outweigh the risks, if any, that may be present.

TABLE 19-5 Variation of Ultrasound Attenuation Coefficient α with Frequency in Megahertz, Where α_1 Is the Attenuation Coefficient at 1 MHz

Tissue	*Frequency Variation*	*Material*	*Frequency Variation*
Blood	$\alpha = \alpha_1 \times \nu$	Lung	$\alpha = \alpha_1 \times \nu$
Fat	$\alpha = \alpha_1 \times \nu$	Liver	$\alpha = \alpha_1 \times \nu$
Muscle (across fibers)	$\alpha = \alpha_1 \times \nu$	Brain	$\alpha = \alpha_1 \times \nu$
Muscle (along fibers)	$\alpha = \alpha_1 \times \nu$	Kidney	$\alpha = \alpha_1 \times \nu$
Aqueous and vitreous		Spinal cord	$\alpha = \alpha^1 \times \nu$
humor of eye	$\alpha = \alpha_1 \times \nu$	Water	$\alpha = \alpha_1 \times \nu^2$
Lens of eye	$\alpha = \alpha_1 \times \nu$	Caster oil	$\alpha = \alpha_1 \times \nu^2$
Skull bone	$\alpha = \alpha_1 \times \nu^2$	Lucite	$\alpha = \alpha_1 \times \nu$

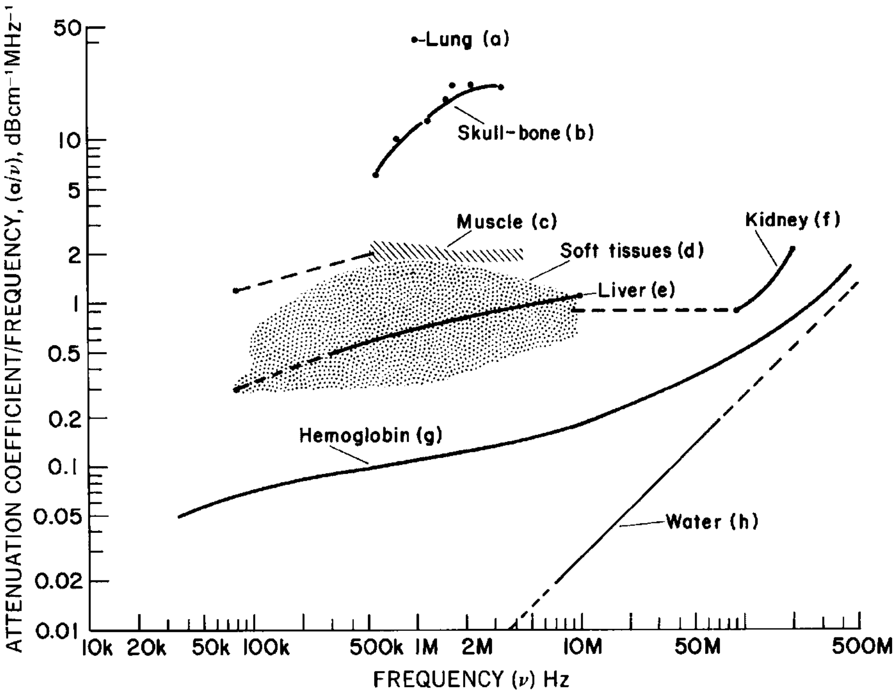


FIGURE 19-6
Ultrasound attenuation coefficient as a function of frequency for various tissue samples. (From Wells, P. N. T.[18] Used with permission.)

If gas bubbles are present in a material through which a sound wave is passing, the compressions and rarefactions cause the bubble to shrink and expand in resonance with the sound wave. The oscillation of such bubbles is referred to as *stable cavitation*. Stable cavitation is not a major mechanism for absorption at ultrasound intensities used diagnostically, but it can be a significant source of scatter.

If an ultrasound beam is intense enough and of the right frequency, the ultrasound-induced mechanical disturbance of the medium can be so great that microscopic bubbles are produced in the medium. The bubbles are formed at foci, such as molecules in the rarefaction zones, and may grow to a cubic millimeter or so in size. As the pressure in the rarefaction zone increases during the next phase of the ultrasound cycle, the bubbles shrink to 10^{-2} mm^3 or so and collapse, thereby creating minute shock waves that seriously disturb the medium if produced in large quantities. The effect, termed *dynamic cavitation*, produces high temperatures (up to 10,000°C) at the point where the collapse occurs.[19] Dynamic cavitation is associated with absorption of energy from the beam. Free radicals are also produced in water surrounding the collapse. Dynamic cavitation is not a significant mechanism of attenuation at diagnostic intensities, although there is evidence that it may occur under certain conditions.[20]

Dynamic cavitation is also termed "transient cavitation."

SUMMARY

- Properties of ultrasound waves include:
  - Compression and rarefaction
  - Requires a transmissive medium
  - Constructive and destructive interference
- The relative intensity and pressure of ultrasound waves are described in units of decibels.

- Ultrasound may be reflected or refracted at a boundary between two media. These properties are determined by the angle of incidence of the ultrasound and the impedance mismatch at the boundary.
- Energy may be removed from an ultrasound beam by various processes, including relaxation energy loss.
- The presence of gas bubbles in a medium may give rise to stable and dynamic cavitation.

PROBLEMS

19-1. Explain what is meant by a longitudinal wave, and describe how an ultrasound wave is propagated through a medium.

*19-2. An ultrasound beam is attenuated by a factor of 20 in passing through a medium. What is the attenuation of the medium in decibels?

*19-3. Determine the fraction of ultrasound energy transmitted and reflected at interfaces between (a) fat and muscle and (b) lens and aqueous and vitreous humor of the eye.

*19-4. What is the angle of refraction for an ultrasound beam incident at an angle of 15 degrees from muscle into bone?

19-5. Explain why refraction contributes to resolution loss in ultrasound imaging.

*19-6. A region of tissue consists of 3 cm fat, 2 cm muscle (ultrasound propagated parallel to fibers), and 3 cm liver. What is the approximate total energy loss of ultrasound in the tissue?

*For those problems marked with an asterisk, answers are provided on p. 493.

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CHAPTER

20

ULTRASOUND TRANSDUCERS

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ISBN: 0-471-38226-4

OBJECTIVES

After studying this chapter, the reader should be able to:

- Explain the piezoelectric effect and its use in ultrasound transducers.
- Characterize the properties of an ultrasound transducer, including those that influence the resonance frequency.
- Describe the properties of an ultrasound beam, including the Fresnel and Fraunhofer zones.
- Delineate the characteristics of focused ultrasound beams and various ultrasound probes.
- Identify different approaches to multitransducer arrays and the advantages of each.

INTRODUCTION

A transducer is any device that converts one form of energy into another. An ultrasound transducer converts electrical energy into ultrasound energy and vice versa. Transducers for ultrasound imaging consist of one or more piezoelectric crystals or elements. The basic properties of ultrasound transducers (resonance, frequency response, focusing, etc.) can be illustrated in terms of single-element transducers. However, imaging is often preformed with multiple-element "arrays" of piezoelectric crystals.

PIEZOELECTRIC EFFECT

The *piezoelectric effect* is exhibited by certain crystals that, in response to applied pressure, develop a voltage across opposite surfaces.[1–3] This effect is used to produce an electrical signal in response to incident ultrasound waves. The magnitude of the electrical signal varies directly with the wave pressure of the incident ultrasound. Similarly, application of a voltage across the crystal causes deformation of the crystal—either compression or extension depending upon the polarity of the voltage. This deforming effect, termed the *converse piezoelectric effect,* is used to produce an ultrasound beam from a transducer.

The piezoelectric effect was first described by Pierre and Jacques Curie in 1880.

The movement of the surface of a piezoelectric crystal used in diagnostic imaging is on the order of a few micrometers (10^{-3} mm) at a rate of several million times per second. This movement, although not discernible to the naked eye, is sufficient to transmit ultrasound energy into the patient.

An ultrasound transducer driven by a continuous alternating voltage produces a continuous ultrasound wave. Continuous-wave (CW) transducers are used in CW Doppler ultrasound. A transducer driven by a pulsed alternating voltage produces ultrasound bursts that are referred to collectively as pulse-wave ultrasound. Pulsed ultrasound is used in most applications of ultrasound imaging. Pulsed Doppler uses ultrasound pulses of longer duration than those employed in pulse-wave imaging.

Many crystals exhibit the piezoelectric effect at low temperatures, but are unsuitable as ultrasound transducers because their piezoelectric properties do not exist at room temperature. The temperature above which a crystal's piezoelectric properties disappear is known as the *Curie point* of the crystal.

A common definition of the efficiency of a transducer is the fraction of applied energy that is converted to the desired energy mode. For an ultrasound transducer, this definition of efficiency is described as the electromechanical coupling coefficient k_c. If mechanical energy (i.e., pressure) is applied, we obtain

$$k_c^2 = \frac{\text{Mechanical energy converted to electrical energy}}{\text{Applied mechanical energy}}$$

If electrical energy is applied, we obtain

$$k_c^2 = \frac{\text{Electrical energy converted to mechanical energy}}{\text{Applied electrical energy}}$$

Values of k_c for selected piezoelectric crystals are listed in Table 20-1.

Essentially all diagnostic ultrasound units use piezoelectric crystals for the generation and detection of ultrasound. A number of piezoelectric crystals occur in nature (e.g., quartz, Rochelle salts, lithium sulfate, tourmaline, and ammonium dihydrogen phosphate [ADP]). However, crystals used clinically are almost invariable man-made

TABLE 20-1 Properties of Selected Piezoelectric Crystals

Materials	*Electromechanical Coupling Coefficient (K_c)*	*Curie Point (°C)*
Quartz	0.11	550
Rochelle salt	0.78	45
Barium titanate	0.30	120
Lead zirconate titanate (PZT-4)	0.70	328
Lead zirconate titanate (PZT-5)	0.70	365

ceramic ferroelectrics. The most common man-made crystals are barium titanate, lead metaniobate, and lead zirconate titanate (PZT).

In some transducers of newer design, the piezoelectric ceramic is mixed with epoxy to form a *composite ceramic*. Composite ceramics have several performance advantages in comparison with conventional ceramics.[4]

TRANSDUCER DESIGN

The piezoelectric crystal is the functional component of an ultrasound transducer. A crystal exhibits its greatest response at the *resonance frequency*. The resonance frequency is determined by the thickness of the crystal (the dimension of the crystal along the axis of the ultrasound beam). As the crystal goes through a complete cycle from contraction to expansion to the next contraction, compression waves move toward the center of the crystal from opposite crystal faces. If the crystal thickness equals one wavelength of the sound waves, the compressions arrive at the opposite faces just as the next crystal contraction begins. The compression waves oppose the contraction and "dampen" the crystal's response. Therefore it is difficult (i.e., energy would be wasted) to "drive" a crystal with a thickness of one wavelength. If the crystal thickness equals half of the wavelength, a compression wave reaches the opposite crystal face just as expansion is beginning to occur. Each compression wave produced in the contraction phase aids in the expansion phase of the cycle. A similar result is obtained for any odd multiple of half wavelengths (e.g., $3\lambda/2$, $5\lambda/2$), with the crystal progressing through more than one cycle before a given compression wave arrives at the opposite face. Additional crystal thickness produces more attenuation, so the most efficient operation is achieved for a crystal with a thickness equal to half the wavelength of the desired ultrasound. A crystal of half-wavelength thickness resonates at a frequency ν:

The components of an ultrasound transducer include the

- Piezoelectric crystal
- Damping material
- Electrodes
- Housing
- Matching layer
- Insulating cover

$$\nu = \frac{c}{\lambda} = \frac{c}{2t}$$

where $\lambda = 2t$

The transducer described here is a "single-element" transducer. Such a transducer is used in some ophthalmological, m-mode, and pulsed Doppler applications. Most other applications of ultrasound employ one of the multielement transducers described later in this chapter.

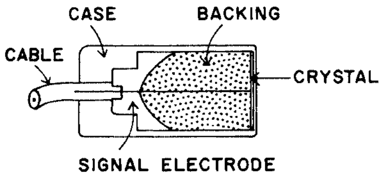
Example 20-1

For a 1.5-mm-thick quartz disk (velocity of ultrasound in quartz = 5740 m/sec), what is the resonance frequency?

$$\nu = \frac{5740 \text{ m/sec}}{2(0.0015 \text{ m})} = 1.91 \text{ MHz}$$

High-frequency ultrasound transducers employ thin (<1 mm) piezoelectric crystals. A thicker crystal yields ultrasound of lower frequency.

To establish electrical contract with a piezoelectric crystal, faces of the crystal are coated with a thin conducting film, and electric contacts are applied. The crystal is mounted at one end of a hollow metal or metal-lined plastic cylinder, with the front face of the crystal coated with a protective plastic that provides efficient transfer of sound between the crystal and the body. The plastic coating at the face of the crystal has a thickness of $\frac{1}{4}\lambda$ and is called a *quarter-wavelength* matching layer. A $\frac{1}{4}\lambda$ thickness maximizes energy transfer from the transducer to the patient. An odd multiple of one-quarter wavelengths would perform the same function, but the greater thickness of material would increase attenuation. Therefore, a single one-quarter wavelength thickness commonly is used for the matching layer. The front face of the crystal is connected through the cylinder to ground potential. The remainder of the crystal is electrically and acoustically insulated from the cylinder.



MARGIN FIGURE 20-1
Typical ultrasound transducer.

With only air behind the crystal, ultrasound transmitted into the cylinder from the crystal is reflected from the cylinder's opposite end. The reflected ultrasound reinforces the ultrasound propagated in the forward direction from the transducer. This reverberation of ultrasound in the transducer itself contributes energy to the ultrasound beam. It also extends the pulse duration (the time over which the ultrasound pulse is produced). Extension of the pulse duration (sometimes called temporal pulse length) is no problem in some clinical uses of ultrasound such as continuous-wave and pulsed-Doppler applications. For these purposes, ultrasound probes with air-backed crystals may be used. However, most ultrasound imaging applications utilize short pulses of ultrasound, and suppression of ultrasound reverberation in the transducer is desirable. Suppression or "damping" of reverberation is accomplished by filling the transducer cylinder with a backing material such as tungsten powder embedded in epoxy resin. Sometimes, rubber is added to the backing to increase the absorption of ultrasound. Often, the rear surface of the backing material is sloped to prevent direct reflection of ultrasound pulses back to the crystal. The construction of a typical ultrasound transducer is illustrated in the margin. The crystal may be flat, as shown in the drawing, or curved to focus the ultrasound beam.

Damping the reverberation of an ultrasound transducer is similar to packing foam rubber around a ringing bell.

The few reverberation cycles after each voltage pulse applied to a damped ultrasound crystal is described as the *ringdown* of the crystal.

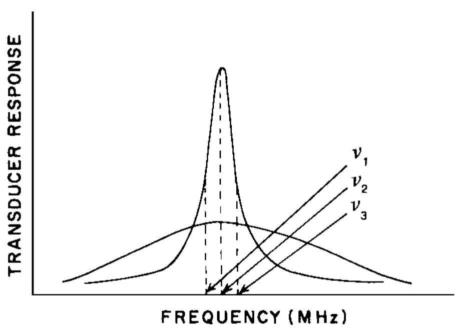
As a substitute for physical damping with selected materials placed behind the crystal, electronic damping may be used. In certain applications, including those that use a small receiving transducer, a resistor connecting the two faces of an air-backed crystal may provide adequate damping. Another approach, termed *dynamic damping,* uses an initial electrical pulse to stimulate the transducer, followed immediately by a voltage pulse of opposite polarity to suppress continuation of transducer action.

FREQUENCY RESPONSE OF TRANSDUCERS

Ultrasound frequencies from 2–10 MHz are used for most diagnostic medical applications.

An ultrasound transducer is designed to be maximally sensitive to ultrasound of a particular frequency, termed the *resonance frequency* of the transducer. The resonance frequency is determined principally by the thickness of the piezoelectric crystal. Thin crystals yield high resonance frequencies, and vice versa. The resonance frequency is revealed by a curve of transducer response plotted as a function of ultrasound frequency. In the illustration in the margin, the frequency response characteristics of two transducers are illustrated. The curve for the undamped transducer displays a sharp frequency response over a limited frequency range. Because of greater energy absorption in the damped transducer, the frequency response is much broader and not so sharply peaked at the transducer resonance frequency. On the curve for the undamped transducer, points ν_1 and ν_3 represent frequencies on either side of the resonance frequency where the response has diminished to half. These points are called the half-power points, and they encompass a range of frequencies termed the bandwidth of the transducer. The ratio of the "center," or resonance frequency ν_2, to the bandwidth $(\nu_3 - \nu_t)$ is termed the Q value of the transducer. The Q value describes the sharpness of the frequency response curve, with a high Q value indicating a sharply-peaked frequency response.

$$Q \text{ value} = \frac{\nu_2}{\nu_3 - \nu_1}$$



MARGIN FIGURE 20-2
Frequency–response curves for undamped (*sharp curve*) and damped (*broad curve*) transducers.

Transducers used in ultrasound imaging must furnish short ultrasound pulses and respond to returning echoes over a wide range of frequencies. For these reasons, heavily damped transducers with low Q values (e.g., 2 to 3) are usually desired.

Because part of the damping is provided by the crystal itself, crystals such as PZT (lead zirconate titanate) or lead metaniobate with high internal damping and low Q values are generally preferred for imaging.

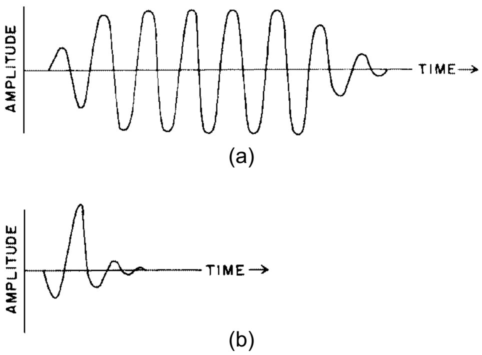
The efficiency with which an ultrasound beam is transmitted from a transducer into a medium (and vice versa) depends on how well the transducer is coupled to the medium. If the acoustic impedance of the coupling medium is not too different from that of either the transducer or the medium and if the thickness of the coupling medium is much less than the ultrasound wavelength, then the ultrasound is transmitted into the medium with little energy loss. Transmission with almost no energy loss is accomplished, for example, with a thin layer of oil placed between transducer and skin during a diagnostic ultrasound examination. Transmission with minimum energy loss occurs when the impedance of the coupling medium is intermediate between the impedances of the crystal and the medium. The ideal impedance of the coupling medium is

$$Z_{\text{coupling medium}} = \sqrt{Z_{\text{transducer}} \times Z_{\text{medium}}}$$

The resonance frequency ν_2 is near the value posted on the imaging system for the transducer. For example, an individual "3.5"-MHz transducer may have an actual resonance frequency of 3.489 MHz.

Coupling of the transducer to the transmitting medium affects the size of the electrical signals generated by returning ultrasound pulses.

Two methods are commonly used to generate ultrasound beams. For continuous-wave beams, an oscillating voltage is applied with a frequency equal to that desired for the ultrasound beam. A similar voltage of prescribed duration is used to generate long pulses of ultrasound energy, as shown in the margin **(A)**. For clinical ultrasound imaging, short pulses usually are preferred. These pulses are produced by shocking the crystal into mechanical oscillation by a momentary change in the voltage across the crystal. The oscillation is damped quickly by the methods described earlier to furnish ultrasound pulses as short as half a cycle. The duration of a pulse usually is defined as the number of half cycles in the pulse with an amplitude greater than one fourth of peak amplitude. The effectiveness of damping is described by the pulse dynamic range, defined as the ratio of the peak amplitude of the pulse divided by the amplitude of ripples following the pulse. A typical ultrasound pulse of short duration is illustrated in the margin **(B)**.



MARGIN FIGURE 20-3
Typical long **(A)** and short **(B)** ultrasound pulses.

ULTRASOUND BEAMS

Wave Fronts

The compression zones of an ultrasound wave are represented by lines perpendicular to the motion of the ultrasound wave in the medium. These lines are referred to as *wave fronts*. For an ultrasound source of large dimensions (i.e., a large-diameter transducer as compared with the wavelength), ultrasound wave fronts are represented as equally spaced straight lines such as those in Figure 20-1A. Wave fronts of this type are termed *planar wave fronts*, and the ultrasound wave they represent is termed a *planar* or *plane wave*. At the other extreme, an ultrasound wave originating from a source of very small dimensions (i.e., a point source) is represented by wave fronts that describe spheres of increasing diameter at increasing distance from the source. *Spherical wave fronts* from a point source are diagramed in Figure 20-1B.

Ultrasound from a point source creates spherical wave fronts. Ultrasound from a two-dimensional extended source creates planar wavefronts.

Sources of exceptionally small or large dimensions are not routinely used in diagnostic ultrasound. Instead, sources with finite dimensions are used. These sources

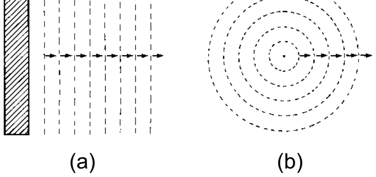


FIGURE 20-1
Ultrasound wave fronts from a source of large dimensions **(A)** and small dimensions **(B)**.

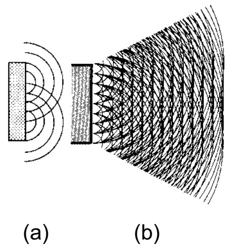


FIGURE 20-2
A: Ultrasound sources may be considered to be a collection of point sources, each radiating spherical wavelets into the medium. **B:** Interference of the spherical wavelets establishes a characteristic pattern for the resulting net wavefronts.

The principles of interference were described by Christian Huygens (1629–1695), the Dutch mathematician, physicist, and astronomer.

can be considered to be a collection of point sources, each radiating spherical wave fronts (termed wavelets) into the medium, as shown in Figure 20-2A. In regions where compression zones for one wavelet intersect those of another, a condition of constructive interference is established. With constructive interference, the wavelets reinforce each other, and the total pressure in the region is the sum of the pressures contributed by each wavelet. In regions where compression zones for one wavelet intersect zones of rarefaction for another wavelet, a condition of destructive interference is established. In these regions, the molecular density is reduced.

Ultrasound transducer rules: (1) The length of the near field increases with increasing transducer diameter and frequency; (2) divergence in the far field decreases with increasing transducer diameter and frequency.

With many spherical wavelets radiating from a transducer of reasonable size (i.e., the diameter of the transducer is considerably larger than the ultrasound wavelength), many regions of constructive and destructive interference are established in the medium. In Figure 20-2B, these regions are represented as intersections of lines depicting compression zones of individual wavelets. In this figure, the reinforcement and cancellation of individual wavelets are most noticeable in the region near the source of ultrasound. They are progressively less dramatic with increasing distance from the ultrasound source. The region near the source where the interference of wavelets is most apparent is termed the *Fresnel* (or *near*) *zone*. For a disk-shaped transducer of radius r, the length D of the Fresnel zone is

The length D of the Fresnel (near) zone can also be written as $D = d^2/4\lambda$, where d is the transducer diameter.

$$D_{\text{fresnel}} = \frac{r^2}{\lambda}$$

where λ is the ultrasound wavelength. Within the Fresnel zone, most of the ultrasound energy is confined to a beam width no greater than the transducer diameter. Beyond the Fresnel zone, some of the energy escapes along the periphery of the beam to produce a gradual divergence of the ultrasound beam that is described by

$$\sin\theta = 0.6\left(\frac{\lambda}{r}\right)$$

where θ is the Fraunhofer divergence angle in degrees (see margin). The region beyond the Fresnel zone is termed *the Fraunhofer* (òr *far*) *zone*.

Rules for Transducer Design
For a given transducer diameter,

- the near-field length increases with increasing frequency;
- beam divergence in the far field decreases with increasing frequency

For a given transducer frequency

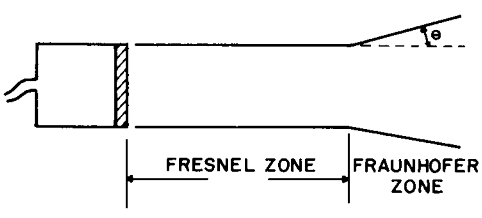
- the near-field length increases with increasing transducer diameter;
- beam divergence in the far field decreases with increasing transducer diameter.

Example 20-2

What is the length of the Fresnel zone for a 10-mm-diameter, 2-MHz unfocused ultrasound transducer?

A 10-mm-diameter transducer has a radius of 5 mm. A 2-MHz transducer has a λ of

$$\begin{aligned}\lambda &= \frac{1540 \text{ m/sec}}{2 \times 10^6/\text{sec}} = 770 \times 10^{-6} \text{ m}\\ &= 0.77 \text{ mm}\\ D_{\text{Fresnel}} &= \frac{(5 \text{ mm})^2}{0.77 \text{ mm}}\\ &= 32.5 \text{ mm}\end{aligned}$$



MARGIN FIGURE 20-4
Divergence of the ultrasound beam in the Fraunhofer region. Angle θ is the Fraunhofer divergence angle.

TABLE 20-2 Transducer Radius and Ultrasound Frequency and Their Relationship to Fresnel Zone Depth and Beam Divergence

Frequency (MHz)	*Wavelength (cm)*	*Fresnel Zone Depth (cm)*	*Fraunhofer Divergence Angle (degrees)*
Transducer radius constant at 0.5 cm			
0.5	0.30	0.82	21.5
1.0	0.15	1.63	10.5
2.0	0.075	3.25	5.2
4.0	0.0325	6.50	2.3
8.0	0.0163	13.0	1.1

Radius (cm)	*Fresnel Zone Depth (cm)*	*Fraunhofer Divergence Angle in Water (degrees)*
Frequency constant at 2 MHz		
0.25	0.83	10.6
0.5	3.33	5.3
1.0	13.33	2.6
2.0	53.33	1.3

For medical applications of ultrasound, beams with little lateral dispersion of energy (i.e., long Fresnel zones) are preferred. Hence a reasonably high ratio of transducer radius to wavelength (r/λ) is desired. This requirement can be satisfied by using ultrasound of short wavelengths (i.e., high frequencies). However, the absorption of ultrasound energy increases at higher frequencies, and frequencies for clinical imaging are limited to 2 to 20 MHz. At these frequencies a transducer radius of 10 mm or more usually furnishes an ultrasound beam with adequate directionality for medical use. The relationship of transducer radius and ultrasound frequency to the depth of the Fresnel zone and the amount of beam divergence is illustrated in Table 20-2.

Beam Profiles

The transmission and reception patterns of an ultrasound transducer are affected by slight variations in the construction and manner of electrical stimulation of the transducer. Hence, the exact shape of an ultrasound beam is difficult to predict, and beam shapes or profiles must be measured for a particular transducer. One approach to the display of ultrasound beam characteristics is a set of pulse-echo response profiles. A profile is obtained by placing an ultrasound reflector some distance from the transducer and scanning the transducer in a direction perpendicular to the axis of the ultrasound beam. During the scan, the amplitude of the signal induced in the transducer by the reflected ultrasound is plotted as a function of the distance between the central axis of the ultrasound beam and the reflector. A pulse-echo response profile is shown in Figure 20-3A, and a set of profiles obtained at different distances from the transducer is shown in Figure 20-3B.

In Figure 20-3A, locations are indicated where the transducer response (amplitude of the reflected signal) decreases to half (−6 dB) of the response when the transducer is aligned with the reflector. The distance between these locations is termed the response width of the transducer at the particular distance (range) from the transducer. If response widths are connected between profiles at different ranges (Figure 20-3C), 6-dB response margins are obtained on each side of the ultrasound beam axis. Similarly, 20-dB response margins may be obtained by connecting the $^1/_{10}$ amplitude (−20 dB) response widths on each side of the beam axis.

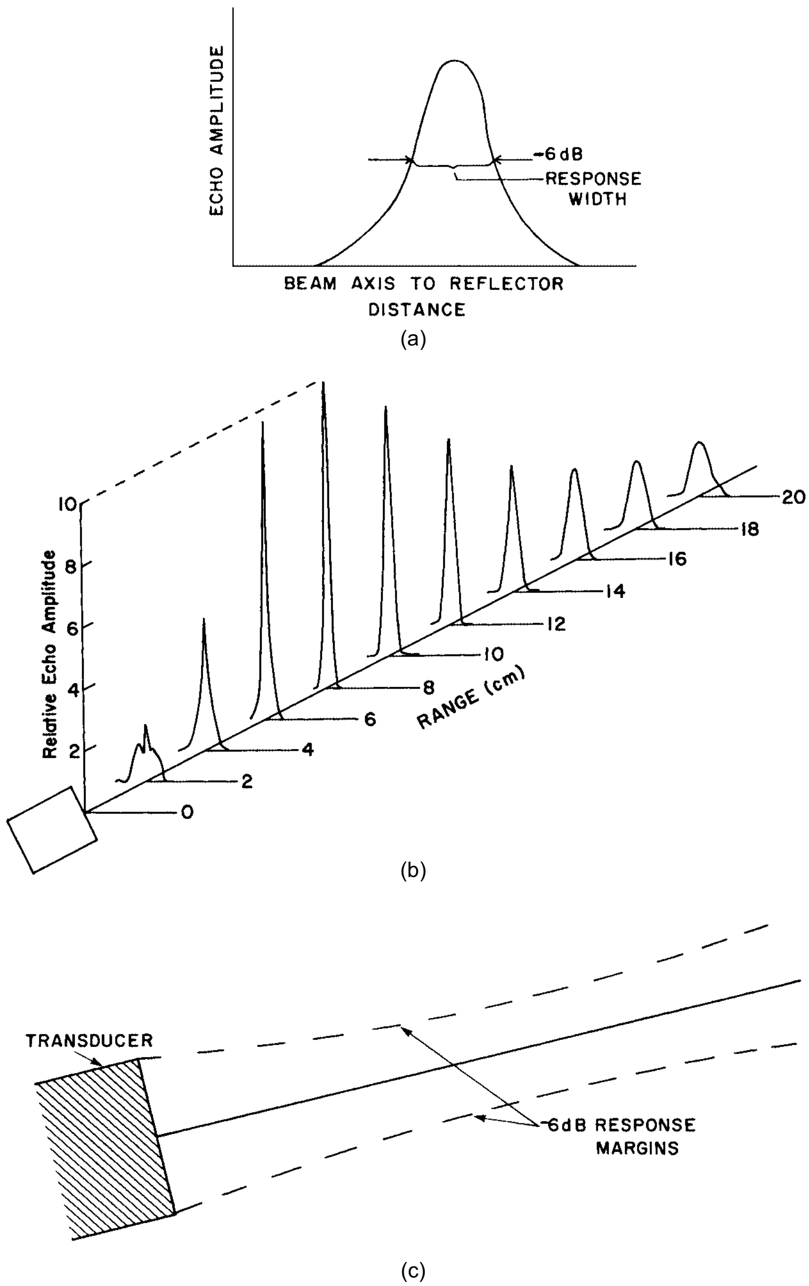
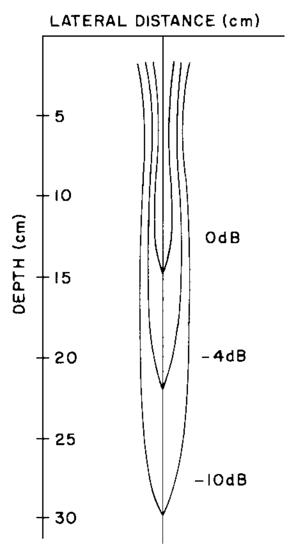


FIGURE 20-3
A: Pulse-echo response profile and the response width of a transducer. **B:** Set of response profiles along the axis of an ultrasound beam. **C:** Response margins of −6 dB along an ultrasound beam.



MARGIN FIGURE 20-5
Isoecho contours for a nonfocused transducer.

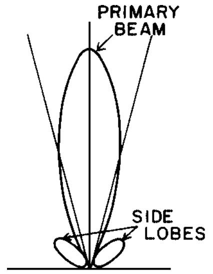
Response profiles for a particular transducer are influenced by several factors, including the nature of the stimulating voltage applied to the transducer, the characteristics of the electronic circuitry of the receiver, and the shape, size, and character of the reflector. Usually, the reflector is a steel sphere or rod with a diameter of three to ten times the ultrasound wavelength. Response profiles may be distorted if the receiver electronics do not faithfully represent low-intensity signals. Some older ultrasound units cannot accurately display echo amplitudes much less than $^1/_{10}$ (−20 dB) of the largest echo recorded. These units are said to have limited dynamic range.

Another approach to describing the character of an ultrasound beam is with isoecho contours. Each contour depicts the locations of equal echo intensity for the ultrasound beam. At each of these locations, a reflecting object will be detected with equal sensitivity. The approach usually used to measure isoecho contours is to place a small steel ball at a variety of positions in the ultrasound beam and to identify locations where the reflected echoes are equal. Connecting these locations with lines yields isoecho contours such as those in the margin, where the region of maximum sensitivity at a particular depth is labeled 0 dB, and isoecho contours of lesser intensity are labeled −4 dB, −10 dB, and so on. Isoecho contours help depict the lateral resolution of a transducer, as well as variations in lateral resolution with depth and with changes in instrument settings such as beam intensity, detector amplifier gain, and echo threshold.

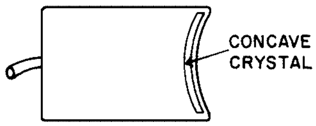
Accompanying a primary ultrasound beam are small beams of greatly reduced intensity that are emitted at angles to the primary beam. These small beams, termed side lobes (see margin), are caused by vibratory modes of the transducer in the transverse plane. Side lobes can produce image artifacts in regions near the transducer, if a particularly echogenic material, such as a biopsy needle, is present.

The preceding discussion covers general-purpose, flat-surfaced transducers. For most ultrasound applications, transducers with special shapes are preferred. Among these special-purpose transducers are focused transducers, double-crystal transducers, ophthalmic probes, intravascular probes, esophageal probes, composite probes, variable-angle probes, and transducer arrays.

Side lobes can be reduced further by the process of *apodization,* in which the voltage applied to the transducer is diminished from the center to the periphery.



MARGIN FIGURE 20-6
Side lobes of an ultrasound beam.

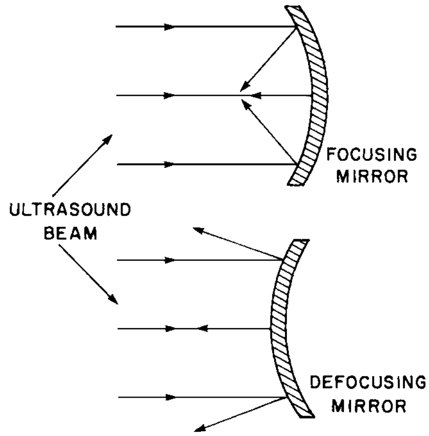


MARGIN FIGURE 20-7
Focused transducer.

Focused Transducers

A focused ultrasound transducer produces a beam that is narrower at some distance from the transducer face than its dimension at the face of the transducer.[5,6] In the region where the beam narrows (termed the focal zone of the transducer), the ultrasound intensity may be heightened by 100 times or more compared with the intensity outside of the focal zone. Because of this increased intensity, a much larger signal will be induced in a transducer from a reflector positioned in the focal zone. The distance between the location for maximum echo in the focal zone and the element responsible for focusing the ultrasound beam is termed the focal length of the transducer.

Often, the focusing element is the piezoelectric crystal itself, which is shaped like a concave disk (see figure in margin). An ultrasound beam also may be focused with mirrors and refracting lenses. Focusing lenses and mirrors are capable of increasing the intensity of an ultrasound beam by factors greater than 100. Focusing mirrors, usually constructed of tungsten-impregnated epoxy resin, are illustrated in the margin. Because the velocity of ultrasound generally is greater in a lens than in the surrounding medium, concave ultrasound lenses are focusing, and convex ultrasound lenses are defocusing (see margin). These effects are the opposite of those for the action of optical lenses on visible light. Ultrasound lenses usually are constructed of epoxy resins and plastics such as polystyrene.

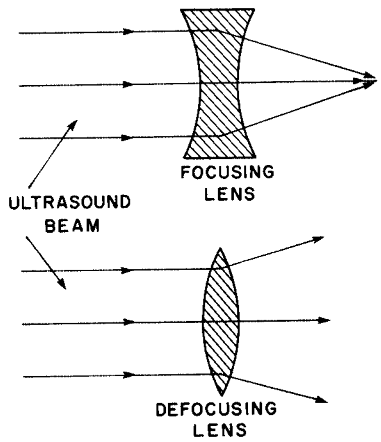


MARGIN FIGURE 20-8
Focusing and defocusing mirrors.

For an ultrasound beam with a circular cross section, focusing characteristics such as pulse-echo response width and relative sensitivity along the beam axis depend on the wavelength of the ultrasound and on the focal length f and radius r of the transducer or other focusing element. These variables may be used to distinguish the degree of focusing of transducers by dividing the near field length r^2/λ by the focal length f. For cupped transducer faces on all but weakly focused transducers, the focal length of the transducer is equal to or slightly shorter than the radius of curvature of the transducer face. If a planoconcave lens with a radius of curvature r is attached to the transducer face, then the focal length f is

$$f = \frac{r}{1 - c_M/c_L}$$

where c_M and c_L are the velocities of ultrasound in the medium and lens, respectively.

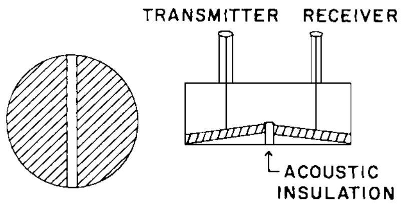


MARGIN FIGURE 20-9
Focusing and defocusing lenses.

The ratio f/d, where $d = 2r$ = the diameter of the transducer, is often described as the f-number of the transducer or other focusing element.

Degree of Focus of Transducers Expressed as a Ratio of the Near-field Length r^2/λ to the Focal Length f

Degree of Focus	*Near–Field Length* / *Focal Length*
Weak	$(r^2/\lambda)/f \leq 1.4$
Medium weak	$1.4 < (r^2/\lambda)/f \leq 6$
Medium	$6 < (r^2/\lambda)/f \leq 20$
Strong	$20 < (r^2/\lambda)/f$



MARGIN FIGURE 20-10
Front (*left*) and side (*right*) views of a typical Doppler transducer.

The length of the focal zone of a particular ultrasound beam is the distance over which a reasonable focus and pulse-echo response are obtained. One estimate of focal zone length is

$$\text{Focal zone length} = 10\lambda\left(\frac{f}{d}\right)^2$$

where $d = 2r$ is the diameter of the transducer. These strongly focused transducers are also used for surgical applications of ultrasound where high ultrasound intensities in localized regions are needed for tissue destruction.

Doppler Probes

Transducers for continuous-wave Doppler ultrasound consist of separate transmitting and receiving crystals, usually oriented at slight angles to each other so that the transmitting and receiving areas intersect at some distance in front of the transducer (see margin). Because a sharp frequency response is desired for a Doppler transducer, only a small amount of damping material is used.

Multiple-Element Transducers

Scanning of the patient may be accomplished by physical motion of a single-element ultrasound transducer. The motion of the transducer may be executed manually by the sonographer or automatically with a mechanical system. Several methods for mechanical scanning are shown in Figure 20-4. The scanning technique that uses an automatic scanning mechanism for the transducer is referred to as *mechanical sector scanning*.

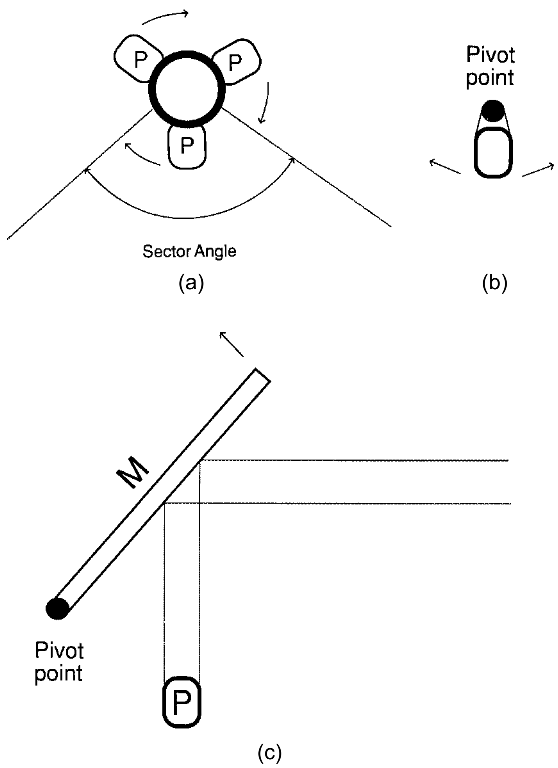


FIGURE 20-4
Three methods of mechanical scanning of the ultrasound beam. **A:** Multiple piezoelectric elements (P) mounted on a rotating head. One element is energized at a time as it rotates through the sector angle. **B:** Single piezoelectric element oscillating at high frequency. **C:** Single piezoelectric element reflected from an oscillating acoustic mirror (M).

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MARGIN FIGURE 20-11
Linear array. (From Zagzebski, J. A.[4] Used with permission.)

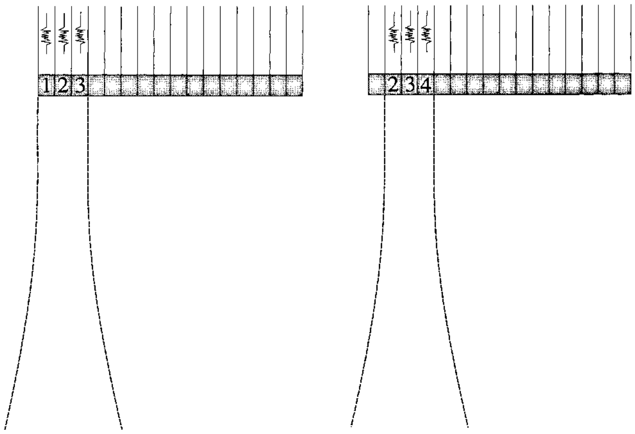

FIGURE 20-5
Electronic scanning with a linear switched array.

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MARGIN FIGURE 20-12
Curved (curvilinear) array. (From Zagzebski, J. A.[4] Used with permission.)

Alternatively, the ultrasound beam may be swept back and forth without the need for any mechanical motion through the use of transducer arrays. Transducer arrays are composed of multiple crystals that can change the direction or degree of focus of the ultrasound beam by timing the excitation of the crystals.

A linear (or curved) array of crystal elements is shown in Figure 20-5. The crystals, as many as 60 to 240 or more, are excited by voltage pulses in groups of three up to 20 or more. Each excitation of the crystal group results in a "scan line" (see next chapter). To obtain the succeeding scan line, the next crystal group is defined to overlap the first (e.g., scan line 1 is produced by crystals 1, 2, and 3, while scan line 2 is produced by crystals 2, 3, and 4, and so on). This pulsing scheme is referred to as a *linear switched array*. By sequentially exciting the entire array, an image composed of a number of scan lines (typically 64, 128, 256, etc.) is obtained.

Another method for scanning with a linear array uses *phased-array* technology. A phase array uses all (typically 128) of the elements of the array to obtain each scan line. By using slight delays between excitations of the elements, the beam may be "swept" to the left or right (Figure 20-6A). A variation in the time delay scheme in a phased array may also be used to focus the beam at various distances (Fig 20-6B) throughout each image. In this technique, called *dynamic focusing,* part of the image is acquired with the focal zone near the transducer face, and part is acquired with the focal zone farther from the transducer face. Thus, two or more images taken at different focal zones may be combined to produce a single image with better axial resolution over a broader range than is possible with a single-element transducer. *Dynamic focusing* is performed without moving parts by simply varying the timing of the excitation pulses.

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MARGIN FIGURE 20-13
Phased array. (From Zagzebski, J. A.[4] Used with permission.)

Advantages of *transducer* arrays:

- They provide electronic beam steering.
- They permit dynamic focusing and beam shaping.

Transducer arrays employ the technique of a *dynamic aperture,* in which the diameter of the sensitive region of the array is expanded as signals are received from greater depths. This technique maintains a constant lateral resolution with depth.

The main distinction between the linear phased array and the linear switched array is that all of the elements of the phased array are used to produce each scan line while only a few of the elements of the switched array are used to produce each scan line. Linear phased arrays provide variable focus in only one dimension—that is, in the plane of the scan. Beam focus in the other dimension, the direction of slice thickness, is provided by acoustic lenses or by concavity of the elements (Figure 20-7).

Another type of array, the annular array, is capable of focusing in all planes perpendicular to the axis of the beam. The annular array (Figure 20-7) consists of a series of piezoelectric elements in the shape of concentric rings or annuli. The beam may be focused at various distances from the transducer face by varying the time delays among excitations of the rings.

Transducer Damage

Ultrasound transducers can be damaged in many ways.[4,7] The crystals are brittle, and the wire contacts on the crystals are fragile; hence transducers should be handled

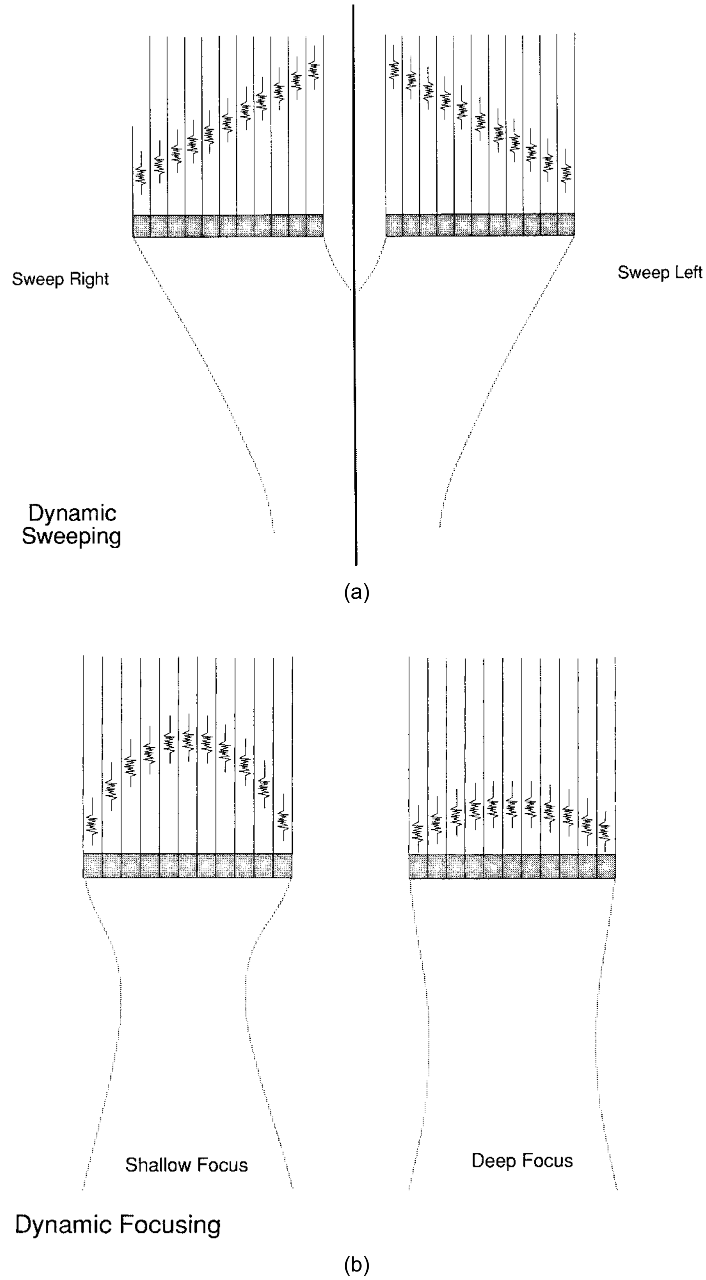

FIGURE 20-6
Electronic scanning with a linear phased array.

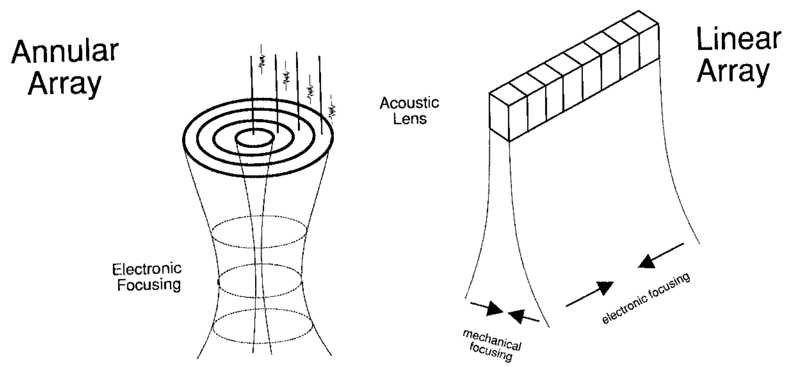

FIGURE 20-7
A linear array requires focusing in the "out-of-plane" dimension by mechanical means (lens or shape of the transducer), whereas an annular array focuses in all directions perpendicular to the axis of the beam.

carefully. Excessive voltages to the crystal should be avoided, and only watertight probes should be immersed in water. A transducer should never be heated to a temperature near the Curie point of the piezoelectric crystal. Dropping a transducer, twisting, bending, and crushing the transducer cables, and other examples of mishandling ultrasound transducers are frequent causes of dysfunction.

PROBLEMS

20-1. Explain what is meant by the piezoelectric effect and the converse piezoelectric effect.

*20-2. For a piezoelectric material with an ultrasound velocity of 6000 m/sec, what thickness should a disk-shaped crystal have to provide an ultrasound beam with a frequency of 2.5 MHz?

20-3. What is meant by damping an ultrasound transducer, and why is this necessary? What influence does damping have on the frequency response of the transducer?

*20-4. What is the estimated focal zone length for a 2-MHz ($\lambda =$ 0.075 cm) focused ultrasound transducer with an f-number of 8?

20-5. A linear array can be electronically steering and focused. An annular array can be electronically focused. Can an annular array be electronically steered?

*For those problems marked with an asterisk, answers are provided on p. 493.

SUMMARY

- The active element of an ultrasound transducer is a crystal that exhibits the piezoelectric (receiver) and converse piezoelectric (transmitter) effect.
- The resonance frequency of a transducer is determined by the type of piezoelectric crystal and its thickness.
- The bandwidth of a transducer is expressed by its Q value.
- The Fresnel zone, the area of minimum beam divergence in front of the transducer, has an axial dimension determined by r^2/λ.
- The angle θ of divergence of the ultrasound beam in the Fraunhofer zone is determined by $\sin\theta = 0.6\,(\lambda/r)$.
- The shape of an ultrasound beam may be described by response profiles or isoecho contours.
- Real-time ultrasound scans may be obtained with mechanical sector scanners or multitransducer arrays.
- Multitransducer arrays include linear switched arrays, linear phased arrays, and annular arrays.

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CHAPTER

21

ULTRASOUND INSTRUMENTATION

Medical Imaging Physics, Fourth Edition, by William R. Hendee and E. Russell Ritenour
ISBN: 0-471-38226-4

OBJECTIVES

After completing this chapter, the reader should be able to:

- Name three modes of ultrasound imaging.
- Determine the time required to obtain an ultrasound image.
- Define pulse repetition frequency and state the role it plays in frame rate.
- Name the main components of an ultrasound imaging system.
- Explain the preprocessing required to obtain digital data from ultrasound signals.
- Define the term *time gain compensation.*
- Explain how postprocessing is used to change the appearance of an ultrasound image.
- Describe several common artifacts in ultrasound images.
- List several common ultrasound quality control tests.

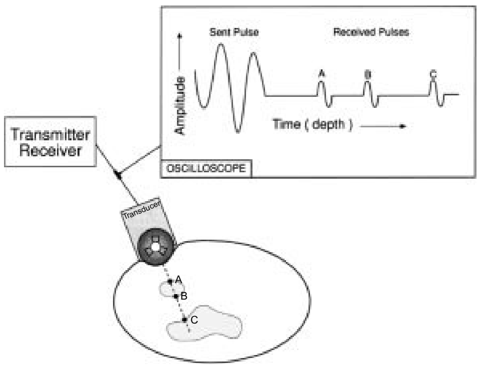
All approaches to obtaining pictorial information with ultrasound depend on echo-ranging, the principle that the time required for return of a reflected signal indicates distance. Strategies for displaying echo-range information—amplitude A, brightness B, and motion M—are discussed in this chapter.

PRESENTATION MODES

A-Mode

In the A-mode presentation of ultrasound images, echoes returning from the body are displayed as signals on an oscilloscope. The oscilloscope presents a graph of voltage representing echo amplitude (hence the term "A-mode") on the ordinate, or y-axis, as a function of time on the abscissa, or x-axis. With the assumption of a constant speed of sound, time on the x-axis can be presented as distance from the ultrasound transducers (Margin Figure 21-1).

A-mode reveals the location of echo-producing structures only in the direction of the ultrasound beam. It has been used in the past to localize echo-producing interfaces such as midline structures in the brain (echoencephalography) and structures to be imaged in B-mode. A-mode displays are not found on most imaging systems used today. The concept of A-mode is however, useful in explaining how pixels are obtained from scan lines in B-mode imaging.



MARGIN FIGURE 21-1
A-mode (amplitude mode) of ultrasound display. An oscilloscope display records the amplitude of echoes as a function of time or depth. Points A, B, and C in the patient appear as peaks-A, B, and C in the A-mode display.

For each centimeter of depth, ultrasound travels 2 cm, 1 cm "out" for the transmitted pulse and 1 cm "back" for the reflected pulse. With the assumption of a speed of sound of 1540 m/sec for ultrasound in soft tissue, the time that corresponds to each centimeter of "depth" is

$$\frac{0.02 \text{ m}}{1540 \text{ m/sec}} = 0.000013\,\text{sec} = 13\,\mu\text{sec}$$

The result is a conversion factor of 13 μsec/cm for diagnostic ultrasound that is used to convert time to distance and vice versa.

This conversion factor is appropriate for reflection imaging in soft tissue where a speed of sound of 1540 m/sec is assumed to be reasonably accurate.

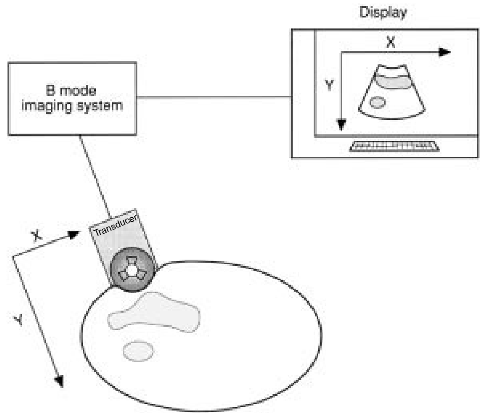
A-mode displays of a transcranial scan (echoencephalography) were used until the late 1970s to detect shifts of the midline in the brain of newborns with suspected hydrocephaly. This technique has been replaced by B-mode real-time scanning, computed tomography and magnetic resonance imaging.

B-Mode

In B-mode presentation of pulse echo images (Margin Figure 21-2) the location of echo-producing interfaces is displayed in two dimensions (x and y) on a video screen. The amplitude of each echo is represented by the brightness value at the xy location. High-amplitude echoes can be presented as either high brightness or low brightness to provide either "white-on-black" or "black-on-white" presentations. Most images are viewed as white on black, so regions in the patient that are more echogenic correspond to regions in the image that are brighter (hence B for "brightness" mode).

B-mode images may be displayed as either "static" or "real-time" images. In static imaging the image is compiled as the sound beam is scanned across the patient, and the image presents a "snapshot" averaged over the time required to sweep the sound beam. In real-time imaging, the image is also built up as the sound beam scans across the patient, but the scanning is performed automatically and quickly, and one image follows another in quick succession. At image frequencies greater than approximately 24 per second, the motion of moving structures seems continuous, even though it may appear that the images are flickering (i.e., being flashed on and off). Images that are refreshed at frequencies greater than approximately 48 per second are free of flicker.

Real-time B-mode images are useful in the display of moving structures such as heart valves. They also permit the technologist to scan through the anatomy to locate structures of interest. Certain applications such as sequential slice imaging of organs are performed better with static imaging.

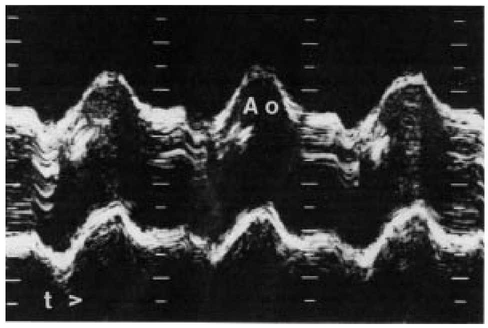


MARGIN FIGURE 21-2
B-mode, or brightness mode, ultrasound display. The amplitude of reflected signals is displayed as brightness at points of an image defined by their x- and y-coordinates.

M-Mode

The M-mode presentation of ultrasound images is designed specifically to depict moving structures. In an M-mode display, the position of each echo-producing interface is presented as a function of time. The most frequent application of M-mode scanning is echocardiography, where the motion of various interfaces in the heart is depicted graphically on a cathode-ray tube (CRT) display or chart recording.

In a typical M-mode display (Margin Figure 21-3), the depths of the structures of interest are portrayed as a series of dots in the vertical direction on the CRT, with the position of the transducer represented by the top of the display. With the transducer in a fixed position, a sweep voltage is applied to the CRT deflection plates to cause the dots to sweep at a controlled rate across the CRT screen. For stationary structures, the dots form horizontal lines in the image. Structures that move in a direction parallel to the ultrasound beam produce vertical fluctuations in the horizontal trace to reveal their motion. The image may be displayed on a short-persistence CRT or storage scope and may be recorded on film or a chart recorder. Modern systems digitize the information and display it on a digital monitor.



MARGIN FIGURE 21-3
A typical M-mode echocardiographic tracing.

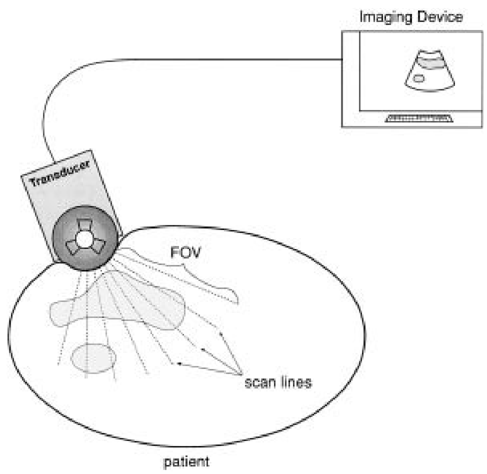
TIME REQUIRED TO OBTAIN IMAGES

Image formation with ultrasound requires that echo information be received along discrete paths called "scan lines." The time required to complete each line is determined by the speed of sound. If all of the information must be received from one line before another can be initiated, a fundamental limit is imposed on the rate at which ultrasound images can be acquired (Margin Figure 21-4).

Each line of information is obtained as follows. First, an ultrasound pulse of several nanoseconds [called the pulse time (PT)] is sent. The transducer is then quiescent for the remainder of the pulse repetition period (PRP), defined as the time from the beginning of one pulse to the next. During the quiescent time, echoes returning from interfaces within the patient excite the transducer and cause voltage pulses to be transmitted to the imaging device. These "echoes" are processed by the device and added to the image only if they fall within a preselected "listen time." Acquisition of echoes during the listen time provides information about reflecting interfaces along a single path in the object—that is, the scan line (Margin Figure 21-5).

The PRP determines the maximum field of view (FOV), also known as depth of view (DOV)—that is, the length of the scan lines.

$$\text{PRP (sec)} = \text{FOV (cm)} \times 13 \times 10^{-6} \text{ sec/cm} \quad \textbf{(21-1)}$$

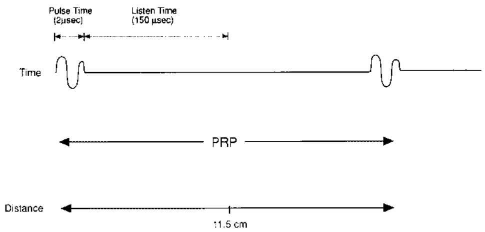


MARGIN FIGURE 21-4
A B-mode image consists of scan lines. The length of the scan lines determines the depth within the patient that is imaged [the field of view (FOV)].

Example 21-1

Find the maximum length of a scan line from an ultrasound unit having a PRP of 195 μs.

$$\begin{aligned}\text{Length of scan line} &= \frac{\text{PRP } (\mu\text{sec})}{13\ \mu\text{sec/cm}}\\ &= \frac{195\ \mu\text{sec}}{13\ \mu\text{sec/cm}}\\ &= 15 \text{ cm}\end{aligned}$$



MARGIN FIGURE 21-5
Ultrasonic echoes from a single scan line in A- or B-mode imaging. In the example shown here, a 150-μsec "listen time" corresponds to a depth [field of view (FOV)] of 11.5 cm. The pulse time (or pulse duration) is short when compared with the listen time. The time from the beginning of one pulse to the beginning of the next is the pulse repetition period (PRP).

The pulse repetition frequency (PRF) is the inverse of the PRP:

$$\text{PRF} = \frac{1}{\text{PRP}}$$

where the PRF has units of pulses per second, inverse seconds, or hertz. The concept of PRF is of special interest in pulsed Doppler ultrasound (Chapter 22).

The frame time (FT) is the time required to obtain a complete image (frame) made up of multiple scan lines (N):

$$\text{FT} = \text{PRP} \times N \qquad \textbf{(21-2)}$$

Obviously, the frame time can be shortened by obtaining two or more scan lines simultaneously. Some linear arrays are capable of providing "parallel" acquisition of image data. For the purpose of problems presented here, we will assume that scan lines are acquired sequentially (i.e., one after the other).

A quantity such as frame time is a "period." It is the time required to complete a periodic task. For any such quantity, there is the inverse, called a "frequency" or "rate," that depicts the number of such tasks that can be performed per unit time. So the rate at which images may be acquired, the frame rate (FR), is the inverse of the frame time.

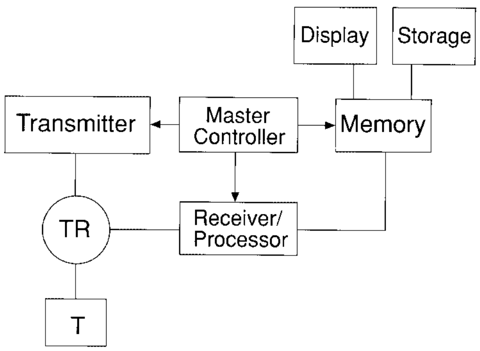
Note that the PRP determines the *maximum* depth of view. The depth of view selected on the imaging device could be smaller than PRP(μs)/13(μs/cm)μsec if the pulse listen time were shorter than the PRP.

$$\text{FR} = \frac{1}{\text{FT}}$$

By substituting from Eqs. (21-1) and (21-2), we obtain

$$\text{FR} = \frac{1}{13 \times 10^{-6} \times \text{FOV} \times N} \qquad \textbf{(21-3)}$$

The frame rate has units of frames per second, inverse seconds, or hertz.



MARGIN FIGURE 21-6
Main components of an ultrasound B-mode imaging system.

Example 21-2

For an ultrasound image with a 10-cm field of view (FOV) and 120 lines of sight, find the minimum pulse repetition period (PRP), the minimum frame time (FT), and the maximum frame rate (FR).

$$\begin{aligned}
\text{PRP} &= \text{FOV} \times 13 \times 10^{-6} \\
&= 10 \text{ cm} \times 13 \times 10^{-6} \text{ sec/cm} \\
&= 130 \times 10^{-6} \\
&= 130\ \mu\text{sec} \\
\text{FT} &= \text{PRP} \times N \\
&= 130 \times 10^{-6} \text{ sec} \times 120 \\
&= 15.6 \times 10^{-3} \text{ sec} \\
&= 15.6 \text{ msec} \\
\text{FR} &= \frac{1}{\text{FT}} \\
&= \frac{1}{15.6 \times 10^{-3} \text{ sec}} \\
&= 64 \text{ sec}^{-1} \\
&= 64 \text{ frames/sec} \\
&= 64 \text{ Hz}
\end{aligned}$$

Ultrasound instrumentation for medical imaging was developed after World War II when personnel, who were familiar with military electronics (signal generators, amplifiers) and the use of sonar, returned to civilian life and wanted to transfer these technologies to medicine. One of the first experimental scanners was built in Denver, CO in 1949 by Joseph Holmes and Douglas Howry. It employed an amplifier from surplus Air Force radar equipment, a power supply from a record player, and a gun turret from a B-29 bomber.

An inverse relationship exists between the frame rate and each of the variables pulse repetition period, field of view, and number of scan lines. An increase in any of these variables decreases the frame rate in direct proportion.

SYSTEM COMPONENTS

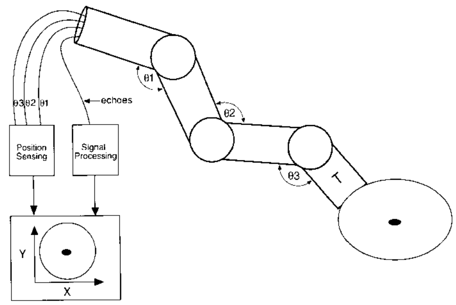
The main components of an ultrasound B-mode imaging system are shown in Margin Figure 21-6. During a scan, the transmitter sends voltage pulses to the transducer. The pulses range from tens to hundreds of volts in amplitude. Returning echoes from the patient produce voltage pulses that are a few microvolts to a volt or so in amplitude. These small signals are sent to the receiver/processor for preprocessing functions such as demodulation, thresholding, amplification, time gain compensation, and detection, all described below. From the receiver/processor, signals are stored as digital values in random-access computer memory.

An automatic transmit/receive switch is provided to prevent large transmitter signals from finding their way into the sensitive receiver/processor. Failure of this switch could result in damage to the receiver/processor, as well as yield no image or one of diminished brightness. The latter condition would call for an increased gain setting beyond that normally used.

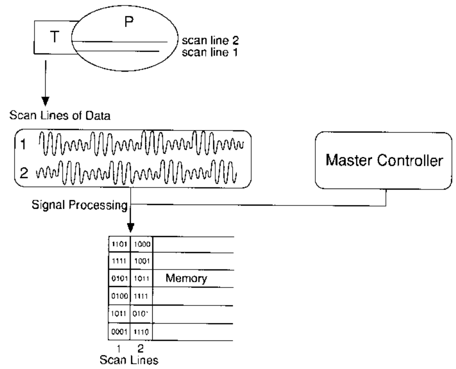
The functions of the components shown in Margin Figure 21-6 are coordinated by the master controller. The master controller provides reference signals against which the various components can time the arrival of echoes or keep track of which scan line is being processed.

For static B-mode imaging, a position-sensing device is added to Margin Figure 21-6. The device is attached to the transducer assembly to monitor the position and orientation of the transducer. Electronic components called "potentiometers" are used in the position-sensing device. When the potentiometer is turned, its resistance changes, producing a change in voltage in a position-sensing circuit. As the position-sensing device is moved, the potentiometers located at hinges provide voltages that are automatically converted to position coordinates (Margin Figure 21-7). For real-time B-mode imaging, the master controller encodes the echoes according to their location in space (Margin Figure 21-8). The echo signals are stored in memory in such a way that when read out as video signals, they appear on the display as brightness values in locations that are meaningful representations of their origins within the patient.

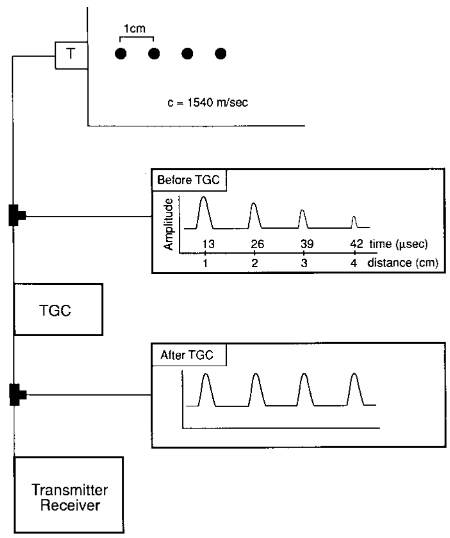
Ultrasound images are usually displayed on a CRT. M-mode data are usually recorded on a strip chart recorder. Static B-mode images usually are recorded on film with a hard-copy camera, and real-time B-mode images usually are preserved on videotape or disk.



MARGIN FIGURE 21-7
The XY position—sensing circuits of a static B-mode scanner. Electric signals from potentiometers 1, 2, and 3 are proportional to the angles θ_1, θ_2, and θ_3 at the hinges of the articulated arm.



MARGIN FIGURE 21-8
In a real-time B-mode imaging system, spatial encoding of the echo signals is accomplished by storing scan-line data in memory in a methodical fashion.



MARGIN FIGURE 21-9
Illustration of the time gain compensation (TGC) principle. Identical steel rods are imaged in a phantom containing an attenuating medium. The time at which echoes return to the transducer (T) depends only upon the speed of sound in the medium. TGC compensates for attenuation by varying the amplification as a function of time of echo reception.

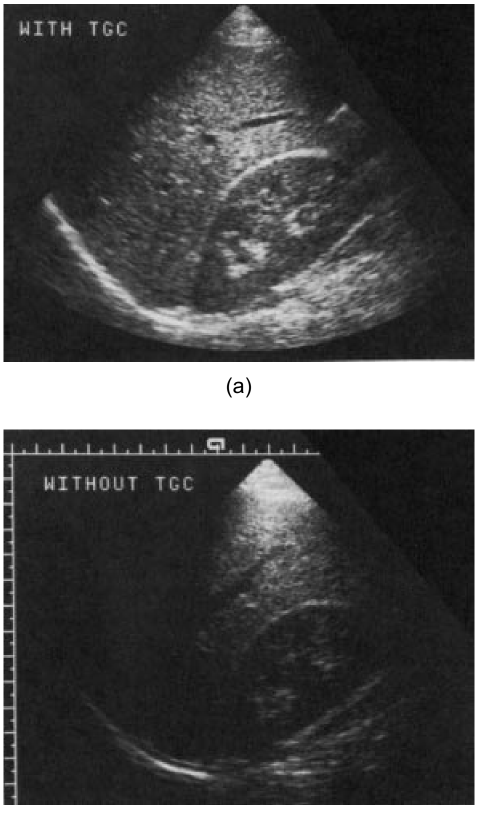
SIGNAL PROCESSING

Returning echoes contain a tremendous amount of information about the patient. As in all imaging modalities, methods used to process the echoes determine the information content of the final image. In B-mode imaging, each location in the image is associated with one value of echo amplitude. This value is recorded in memory and translated into brightness on the display screen. Deriving a single value from each echo is the goal of signal processing.

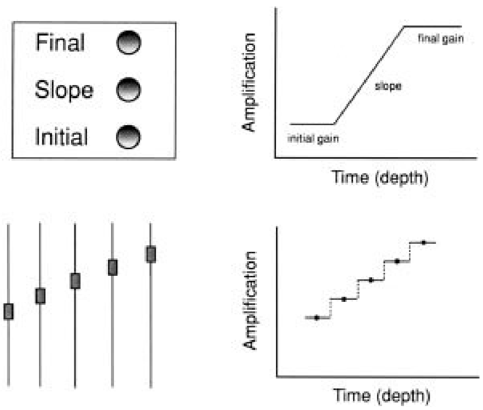
Signal processing may be divided into two categories. If the processing schemes to be used are determined prior to scanning the patient, they are termed *preprocessing schemes*. Postprocessing refers to signal or image processing that alters stored values of data after they have been acquired, (i.e., after the patient has been scanned).

Preprocessing

The amplitude of an ultrasound echo is governed by the acoustic impedance mismatch at the interface where the echoes originated, the attenuation of intervening tissues, and the amplitude of the ultrasonic pulse that is sent out from the transducer. It is the presence and degree of acoustic impedance mismatch at interfaces that we wish to image. The attenuation by intervening tissues is usually considered an undesirable factor because it produces falloff of signal intensity with depth without yielding any useful information. This attenuation can be compensated for in the image by use of the



MARGIN FIGURE 21-10
Real-time B-mode images obtained with **(A)** and without **(B)** a properly adjusted TGC.



MARGIN FIGURE 21-11
Three-knob TGC control **(A)** and a linear potentiometer TGC control **(B)** along with the TGC curves produced.

time gain compensation (TGC) circuit, sometimes called the depth gain compensation (DGC) or time-varied gain (TVG) circuit.

The action of the TGC circuit is shown in Margin Figure 21-9. A plot of uncorrected or "raw" amplitudes or echoes (in the form of an A mode display) from a succession of equivalent reflectors in a homogeneous medium shows a predictable decrease in signal amplitude with depth. The key to compensating for this falloff is the relationship between the depth and time of arrival of the echo at the transducer. Echoes that return at longer times (and therefore from greater depth) receive greater amplification to compensate for increased attenuation by intervening soft tissue. The curve that describes the amplification or gain required to compensate for the attenuation is called the TGC curve. Margin Figure 21-10 shows images obtained with and without a properly adjusted TGC.

The attenuation of different body parts and variations from one patient to another render a single factory setting of the TGC impractical. The TGC must be adjusted for each patient and readjusted during the scan as different tissues are encountered. In practice, most ultrasound systems do not allow specification of a continuous curve to describe the TGC. Instead, a few controls allow reasonable flexibility in determining the curve. Two common types of TGC controls are (1) a series of linear potentiometers for discrete setting of time delays at various depths and (2) a three-knob control that allows adjustment of initial gain, slope, and far gain. The two types of TGC systems are illustrated in Margin Figure 21-11.

Other types of signal preprocessing commonly performed in ultrasound imaging include rejection, rectification, enveloping, and classification (Margin Figure 21-12). Rejection eliminates signals that are too large or too small to be of diagnostic value. All remaining signals are rectified and enveloped, a process sometimes referred to as "low-pass" filtering. Echo intensity may then be defined in a number of ways. Three typical methods are to classify echoes according to (1) the height of the peak value, (2) the area under the peak, or (3) the maximum rate of rise or slope of the echo.

A final preprocessing step involves assigning a value in computer memory that corresponds to signal intensity. Margin Figure 21-13 shows a preprocessing "map" that demonstrates how signal levels measured in decibels are converted into arbitrary numbers to be stored in memory. If the map is linear, for example, then a 10% increase in signal level would be stored as a 10% larger value in memory. By using a nonlinear map, it is possible to accentuate part of the range of echoes at the expense of another. In Margin Figure 21-13 the larger signals, from about 15 to 30 dB, are mapped into stored values of 50 to 256, while smaller signals (0 to 15 dB) are mapped into values of 0 to 50. Thus, the upper half of the range of echoes is accentuated because it takes up four-fifths of the stored values. Relationships among the stronger signals will be accentuated because the gray levels in the display will be taken up disproportionately by the stronger signals. Smaller signals will all appear very dark, with little differentiation. Therefore, the map illustrated in Margin Figure 21-13 provides high signal separation.

Postprocessing

The value stored for each location in the image (each picture element, or pixel) is eventually displayed on a monitor as a level of brightness. That is, the echo intensities are displayed as brightness values varying from black to white. As with preprocessing, one may choose a "map" for postprocessing that may be either linear or nonlinear. A nonlinear map can be designed to emphasize some parts of the range of stored values. In Margin Figure 21-14 a postprocessing map is shown that maps the lower stored values into most of the gray scale, thereby emphasizing differences among the lower values. Note that if the image were preprocessed with the map shown in Margin Figure 21-13 and then postprocessed with the map in Margin Figure 21-14, the result would be the same as if a linear map were used for both. That is, postprocessing can either "undo" or enhance the effects of preprocessing.

Other types of postprocessing features that may be present in an ultrasound device include zoom, region of interest, and digital filtering (see Chapter 10).

DYNAMIC RANGE

The ratio of the largest to the smallest echoes processed by components of an ultrasound device is known as the dynamic range of the device. In general, the dynamic range decreases as signals pass through the imaging system (Margin Figure 21-15) because operations such as TGC and rejection eliminate small and large signals. Echoes returning from tissues can have dynamic ranges of 100 to 150 dB. The dynamic range in decibels may be easily converted to a ratio of amplitudes or intensities.

Example 21-3

Find the amplitude and intensity ratios corresponding to a dynamic range of 100 dB. For amplitudes,

$$\mathrm{dB} = 20 \log \frac{A}{A_0}$$

can be rewritten as

$$\frac{\mathrm{dB}}{20} = \log \frac{A}{A_0}$$

or

$$10^{\mathrm{dB}/20} = \frac{A}{A_0}$$

So, for 100 dB we have

$$10^5 = 100{,}000 = \frac{A}{A_0}$$

The strongest echoes have 100,000 times the amplitude of the weakest echoes. Similarly, for intensity,

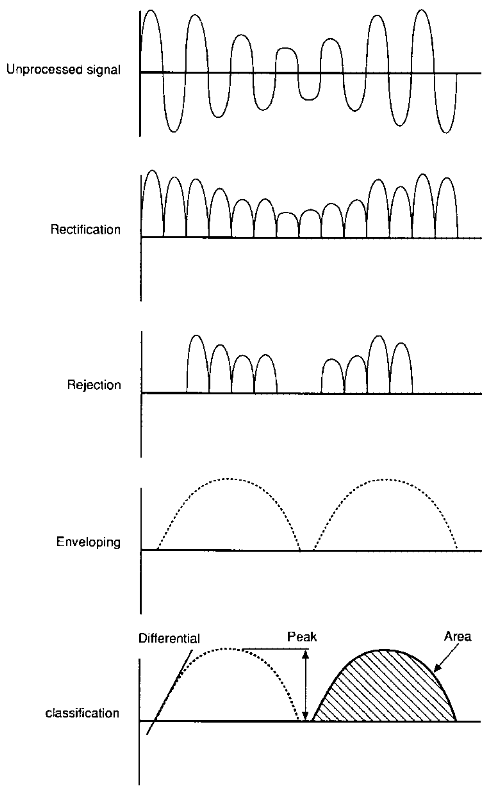
$$\mathrm{dB} = 10 \log \frac{I}{I_0}$$

so that

$$10^{\mathrm{dB}/10} = 10^{10} = 10{,}000{,}000{,}000 = \frac{I}{I_0}$$

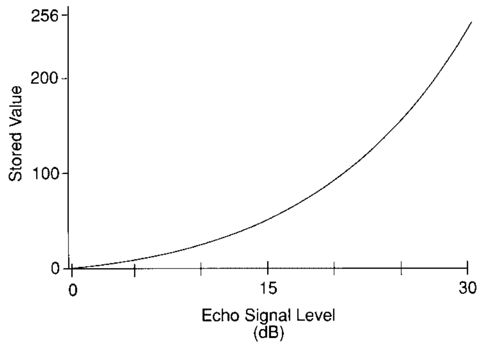
The strongest echoes have 10^{10}, or 10 billion, times the intensity of the weakest echoes.

Some of the dynamic range of echoes is caused by attenuation as the signals traverse several centimeters of tissue. The TGC circuit amplifies the weak echoes that arrive later, and rejection circuits eliminate low-amplitude noise. In this manner, the dynamic range is reduced, or "compressed," to about 40 dB in the receiver/processor. This dynamic range may be retained in memory while a range of about 20 dB is displayed on the system monitor.

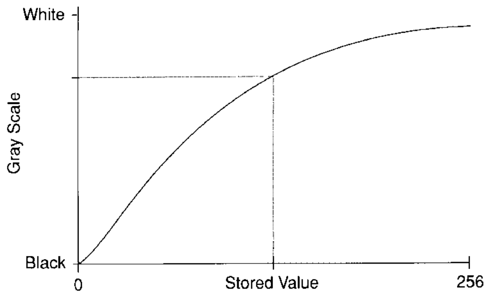


MARGIN FIGURE 21-12
Signal preprocessing in ultrasound.

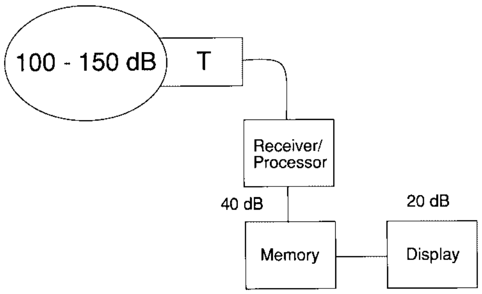
The terms "enveloping" and "low-pass filtering" are used interchangeably in signal processing. A low-pass filter removes higher frequencies but allows lower ones to pass through the circuit. If a complex signal (one consisting of varying amounts of many different frequencies) is sent through a low-pass filter, the resulting signal will look something like the outline or envelope of the original signal.



MARGIN FIGURE 21-13
Preprocessing "map." The graph shows how echo signal levels are converted to stored values. This particular example shows "high signal separation," with the upper half of the amplitude range taking up the upper 4/5 of the stored values.



MARGIN FIGURE 21-14
Postprocessing "map." The graph shows how stored values from memory are converted to gray-scale values. This particular example shows "low signal separation," with the lower half of the stored values taking up the lower 4/5 of the gray scale (brightness values vary from black to white).



MARGIN FIGURE 21-15
Dynamic range of signals at different stages of the ultrasound imaging system.

MARGIN FIGURE 21-16
Acoustic shadowing in an ultrasound image. Stones in a gallbladder reduce the transmission of sound. The display records lower amplitude echoes or "shadows" beyond the region of the stones.

ULTRASOUND IMAGE ARTIFACTS

Even though ultrasound systems produce digital images, it is impossible to identify pixel values in typical clinical images with specific fundamental properties of tissue in the way that x-ray CT numbers are associated with linear attenuation coefficients. Although pixel values in ultrasound images are influenced by fundamental properties such as acoustic impedance and physical density, each pixel value is determined by interrelationships among surrounding materials that are too complex to unfold quantitatively. However, these interrelationships sometimes create artifacts that are recognizable in a qualitative sense.

Speckle

The physical principles behind the phenomenon of speckle were described in Chapter 20. Although the presence of speckle in an image indicates the presence of features in the patient that are smaller than the wavelength of the ultrasound, there is no one-to-one correspondence between a dot in a speckle pattern and the location of a feature. The speckle pattern changes from frame to frame, even for stationary objects, and therefore is not a simple indicator of even the number of such objects. However, various attempts have been made to determine the presence or absence of disease in organs from statistical analysis of speckle patterns.[1]

Shadowing and Enhancement

If some object within the patient has a larger attenuation coefficient than the material that lies beyond it, then the settings of the TGC circuit that would provide appropriate compensation for normal tissue will undercompensate and cause the region beyond the object to appear less echogenic. This phenomenon is referred to as acoustic shadowing (Margin Figure 21-16). Similarly, if the object in the path of the ultrasound beam has a lower attenuation coefficient than its surroundings, acoustic enhancement may result.

Multiple Pathway

Various types of multiple-pathway artifacts occur in ultrasound images (see Chapter 20). When an echo returns to a transducer, the imaging device assumes that the sound traveled in a straight line following a single reflection from some interface in the patient. The scan converter then places the brightness value at an appropriate location in the image. If the actual path of the echo involved multiple reflections, the echo would take longer to return, and the scan converter would place the interface at a greater depth in the image.

Refraction

Refraction sometimes causes displacement of the sound beam as it crosses tissue boundaries. Because the scan converter assumes that the ultrasound travels in straight lines, refraction causes displacement errors in positioning reflective interfaces in the image. A refraction artifact is illustrated in Margin Figure 21-17.

QUALITY CONTROL

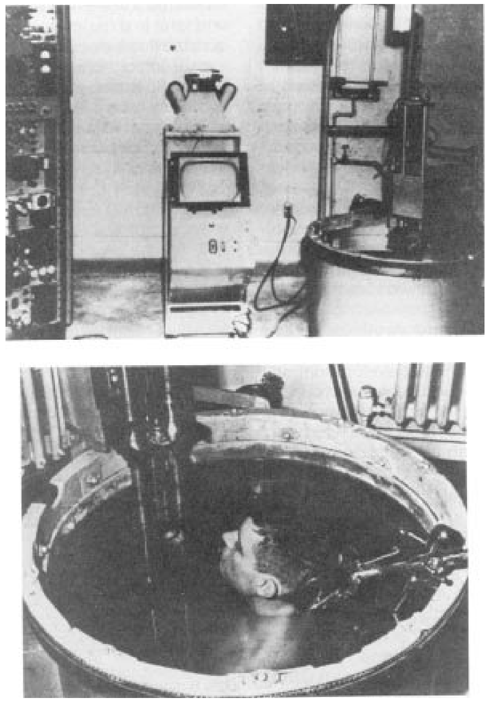
Frequent testing of the performance of the ultrasound imaging system by using phantoms whose properties are well understood can uncover problems that may be too subtle to discern in clinical images.[2] Standard phantoms are shown in Margin

Figure 21-18. Their properties and some protocols for typical performance tests are discussed below.

Shown in Margin Figure 21-18A is an example of a high-contrast test object for evaluation of ultrasound imaging systems. The test object mimics the speed of sound in soft tissue and contains a mixture of ethanol and water or NaCl and water. With this test object, the depth-measuring properties of ultrasound systems can be evaluated. "An equivalent speed of sound" also implies that the wavelength c/ν, near zone length r^2/λ for unfocused transducers, and the focal length for focused transducers are the same in the test medium as in soft tissue. Therefore, this phantom may be used to evaluate axial resolution at an appropriate depth.

The standard American Institute of Ultrasound in Medicine (AIUM) 100-mm test object (Margin Figure 21-18A) is an example of a high-contrast test object that employs a medium with a speed of sound equal to that of soft tissue. Reflected signals are produced by rods having diameters of 0.75 mm located at depths specified to an accuracy of ± 0.25 mm. The diameter and accuracy of location of the rods exceed the specifications for minimum resolution and distance for typical imaging systems.

A tissue-equivalent phantom (Margin Figure 21-18B) mimics more of the ultrasonic properties of human soft tissue as compared with the high-contrast test object. These properties may include scatter characteristics at typical frequencies as well as overall attenuation. The phantom provides an image with a texture pattern similar to that of liver parenchyma with a few low-contrast "cysts" in the phantom.



One of the first ultrasonic scanners for imaging the human body required that the patient be immersed in warm saline solution in a large metal tank in which a transducer rotated about the patient. Head scanning was not often performed with this technique. Shown here is the compound scanner (Somascope) developed by Howry and co-workers from a B-29 gun turret. (From *Am. J. Digest Dis.* 1963; **8**:12. Harper and Row, Hoeber Medical Division. Used with permission.)

Quality Control Tests

Dead Zone

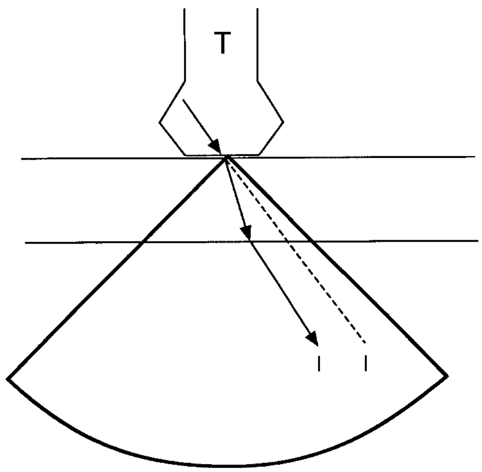
A small portion of the image at the transducer–tissue interface is usually saturated with echoes because of the spatial pulse length of the transducer and multiple reverberations from the transducer–tissue interface. Examination of a group of high-contrast rods located near the surface of the phantom reveals the extent of this problem. The minimum depth at which a high-contrast rod can be resolved is usually quoted as the depth of the dead zone.

Axial Resolution

Any equipment problem that reduces the bandwidth or increases the spatial pulse length of the system interferes with axial resolution. Examples of such problems include cracks in the crystal materials and separation of the facing or backing material. Axial resolution is measured by attempting to resolve (produce a clear image of) rods that are closely spaced along the axis of the sound beam.

Lateral Resolution

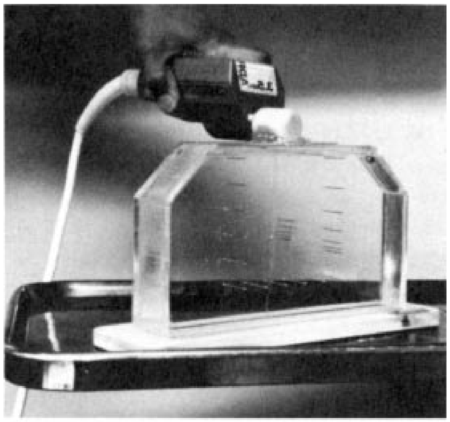
There are several methods to evaluate lateral resolution, defined as resolution in a direction perpendicular to the axis of the ultrasound beam. A scan of rod groups that form a line parallel to the scanning surface can be used. Alternatively, rods that are perpendicular to the axis of the beam may be scanned. Because the lateral resolution *is* the width of the ultrasound beam, the apparent width of the rods scanned in this fashion (Margin Figure 21-19) yields the resolution.



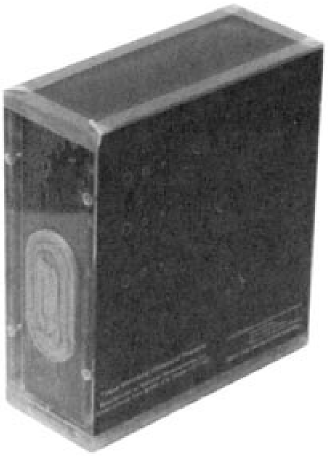
MARGIN FIGURE 21-17
Refraction artifact. The object on the left appears shifted to the right and at a slightly greater depth because of refraction in a layer having lower speed of sound than the rest of the tissue being imaged.

Distance Accuracy Measurement

Distance measurements play a key role in ultrasound dating of fetuses. Errors in distance calibration may lead to a change in the gestational age estimate that would adversely affect patient management. To perform the distance accuracy measurement, high-contrast rods with a separation of at least 10 cm are scanned. Distance



(a)



(b)

MARGIN FIGURE 21-18
Standard ultrasound phantoms. AIUM 100-mm test object **(A)** and a tissue-equivalent phantom **(B)**.

measurement options that are available on the imaging system (marker dots, calipers, etc.) are then used to determine the distance from the image. Because the rod spacings in the test object or phantom are known, the distance error may be reported as a percentage of the true value. For example, a measured distance of 10.5 for a known distance of 10.0 cm is reported as a 5% error. A reasonable goal for most imaging systems currently in use is an error of 2% or less measured over 10 cm.

Sensitivity and Uniformity

There are a number of ways to gauge the ability of an ultrasound system to record faint sources of echoes (sensitivity) and provide a consistent response to echoes (uniformity).[3] Rod groups both perpendicular and parallel to the scanning surface may be scanned, and the brightness and size of each rod may be compared. If a rod group perpendicular to the scanning surface is scanned in an attenuating phantom, then the TGC settings required to render a uniform rod image appearance may be recorded. Images obtained with those TGC settings may be reviewed periodically as a constancy check.

Display and Printing

The consistency of the recording medium may be evaluated with test patterns supplied by the manufacturer. Basically, the appearance of the "hard copy" should be the same as its appearance on the system monitor. This parameter should be measured qualitatively on all systems. More quantitative evaluations involving measurement of monitor luminance are possible[4] but not routine. If a test pattern is not available in vendor-supplied software, it is possible to use any image, such as a typical image of the tissue-equivalent phantom, to establish photographic consistency.

It is most important to separate performance of the image acquisition system from performance of the recording device. This is best accomplished by using a test pattern that is generated independently from the image acquisition system—that is, an "electronic" test pattern. An electronic test pattern that is frequently used for this purpose is a pattern that has been specified by the Society of Motion Picture and Television Engineers (SMPTE). The SMPTE pattern is shown in Margin Figure 21-20.

Frequency of Testing

The quality control tests outlined in this section should be performed monthly and after any service call that may affect the parameters measured.[5] More extensive tests may be performed by qualified personnel upon delivery of the unit.

PROBLEMS

21-1. If A, B, and M mode ultrasound display modes are described as X versus Y, what are the X and Y scales for each mode?

*21-2. How many scan lines may be obtained during a scan of a patient when the depth of view (DOV) is 8 cm if the total scan time is 13.4 msec? Assume sequential acquisition of scan lines.

*21-3. If the pulse repetition period is doubled, the field of view cut in half, and the number of scan lines doubled what is the effect on maximum allowable frame rate? Assume sequential acquisition of scan lines.

21-4. Describe the function of the master controller of a real-time ultrasound B-mode imaging system.

*21-5. What phenomenon is compensated for by the time gain compensation (TGC) circuit?

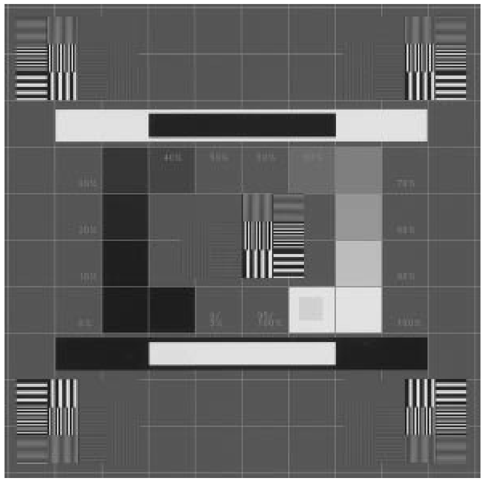
*For those problems marked with an asterisk, answers are provided on p. 493.

SUMMARY

- Display modes include
  - Amplitude
  - Brightness
  - Motion
- Frame rate is inversely proportional to depth of view, number of scan lines, and pulse repetition frequency.
- The main components of an ultrasound imaging system are
  - Transducer
  - Transmit/receive switch
  - Transmitter
  - Receiver/processor
  - Master controller
  - Memory
  - Display
  - Storage
- The time gain compensation system varies the amplification of return signals from each scan line in a variable manner as a function of time after the sent pulse.
- Preprocessing steps include
  - Rectification
  - Pulse height rejection
  - Enveloping (low-pass filtering)
  - Classification (e.g., differential, peak, area, alone or some combination)
- In preprocessing, echo signals are assigned to stored values according to according to a functional form or curve.
- In postprocessing, stored values are assigned to the gray scale according to a functional form or curve.
- Artifacts include
  - Speckle
  - Acoustic shadowing
  - Multiple pathway
  - Refraction
- Ultrasound quality control tests include
  - Dead zone
  - Axial resolution
  - Lateral resolution
  - Distance accuracy
  - Sensitivity and uniformity

(a) (b)

MARGIN FIGURE 21-19
Scan of a rod group in an ultrasound phantom demonstrates beam width (lateral resolution).



MARGIN FIGURE 21-20
The test pattern promulgated by the Society of Motion Pictures and Television Engineers (SMPTE), known colloquially as the "símtē pattern." This pattern, which tests image quality parameters such as high and low contrast resolution, distortion, and linearity, may be used as an electronic data file to test the fidelity of displays, hard-copy devices, or storage and transmission systems. (ref: Society of Motion Pictures and Television Engineers (SMPTE). Specifications for Medical Diagnostic Imaging Test Pattern for Television Monitors and Hard-Copy Recording Cameras. Recommended Practice RP 133-1986. *SMPTE J.* 1986; **95**:693–695.)

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CHAPTER

22

DOPPLER EFFECT

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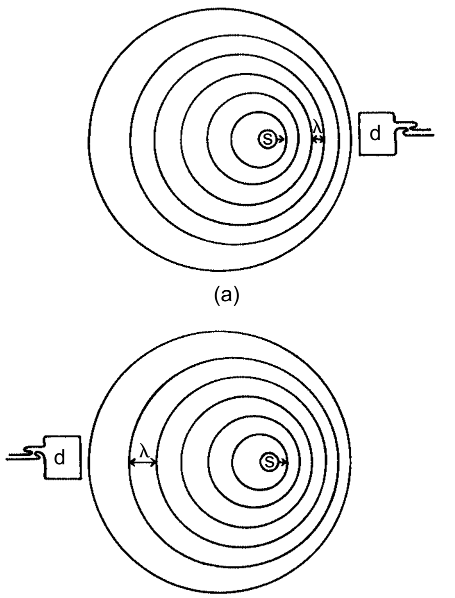


MARGIN FIGURE 22-1
Principles of Doppler ultrasound. **A:** Source moving toward a stationary detector d. **B:** Source moving away from a stationary detector d.

OBJECTIVES

After completing this chapter, the reader should be able to:

- Write the Doppler equation and use it to calculate the Doppler shift.
- Describe the effect of Doppler angle upon frequency shift and describe how error in estimation of the angle influences the accuracy of velocity estimation.
- Determine volumetric flow from average velocity and vessel cross-sectional area.
- Name the main components of a Doppler ultrasound imager.
- Calculate range gate size and depth from timing information.
- Describe the differences between information obtained from pulsed-wave versus continuous-wave Doppler.
- Describe the main features of a frequency spectrum and describe how it is used in constructing the Doppler spectral trace.
- Describe how color is assigned in color Doppler.
- Discuss some reasons for spectral broadening.

Ultrasound is used not only for display of static patient anatomy but also for identification of moving structures in the body. Approaches to the identification of moving structures include real-time pulse-echo imaging, motion mode (M-mode) display of reflected ultrasound pulses, and the Doppler-shift method. Discussed in this chapter are the basic principles of the Doppler-shift method. The Doppler method has a number of applications in clinical medicine, including detection of fetal heartbeat, detection of air emboli, blood pressure monitoring, detection and characterization of blood flow, and localization of blood vessel occlusions.[1,2]

ORIGIN OF DOPPLER SHIFT

When an emergency vehicle with its siren on passes someone on the street, a shift in frequency of one octave on the chromatic musical scale corresponds to a velocity difference between the siren and the listener of 40 miles per hour, two octaves corresponds to 80 miles per hour, and so on.

When there is relative motion between a source and a detector of ultrasound, the frequency of the detected ultrasound differs from that emitted by the source. The shift in frequency is illustrated in Margin Figure 22-1. In Margin Figure 22-1A, an ultrasound source is moving with velocity v_s toward the detector. After time t following the production of any particular wave front, the distance between the wave front and the source is $(c - v_s)t$, where c is the velocity of ultrasound in the medium. The wavelength λ of the ultrasound in the direction of motion is shortened to

$$\lambda = \frac{c - v_s}{\nu_0}$$

where ν_0 is the frequency of ultrasound from the source. With the shortened wavelength, the ultrasound reaches the detector with an increased frequency ν:

$$\nu = \frac{c}{\lambda} = \frac{c}{(c - v_s)/\nu_0}$$
$$= \nu_0\left(\frac{c}{c - v_s}\right)$$

That is, the frequency of the detected ultrasound shifts to a higher value when the ultrasound source is moving toward the detector. The shift in frequency $\Delta\nu$ is

$$\Delta\nu = \nu - \nu_0 = \nu_0\left(\frac{c}{c - v_s}\right) - \nu_0$$
$$= \nu_0\left(\frac{v_s}{c - v_s}\right)$$

If the velocity c of ultrasound in the medium is much greater than the velocity v_s of the ultrasound source, then $c - v_s \simeq c$ and

$$\Delta\nu = \nu_0\left(\frac{v_s}{c}\right)$$

A similar expression is applicable to the case in which the ultrasound source is stationary and the detector is moving toward the source with velocity v_d. In this case the Doppler-shift frequency is approximately

$$\Delta\nu = \nu_0\left(\frac{v_d}{c}\right)$$

where $c >> v_d$.

If the ultrasound source is moving away from the detector (Margin Figure 22-1B), then the distance between the source and a wavefront is $ct + v_s t = (c + v_s)t$, where t is the time elapsed since the production of the wavefront. The wavelength λ of the ultrasound is

$$\lambda = \frac{c + v_s}{\nu_0}$$

and the apparent frequency ν is

$$\begin{aligned}\nu &= \frac{c}{\lambda}\\ &= \frac{c}{(c + v_s)/\nu_0}\\ &= \nu_0\left(\frac{c}{c + v_s}\right)\end{aligned}$$

That is, the frequency shifts to a lower value when the ultrasound source is moving away from the detector. The shift in frequency $\Delta\nu$ is

$$\begin{aligned}\Delta\nu = \nu - \nu_0 &= \nu_0\left(\frac{c}{c + v_s}\right) - \nu_0\\ &= \nu_0\left(\frac{-v_s}{c + v_s}\right)\end{aligned}$$

where the negative sign implies a reduction in frequency. If the velocity c of ultrasound is much greater than the velocity v_s of the source, $c + v_s \simeq c$, and $\Delta\nu = \nu_0(-v_s/c)$. A similar expression is applicable to the case where the ultrasound source is stationary and the detector is moving away from the source with velocity v_d:

$$\Delta\nu = \nu_0\left(\frac{-v_d}{c}\right)$$

when $c >> v_d$.

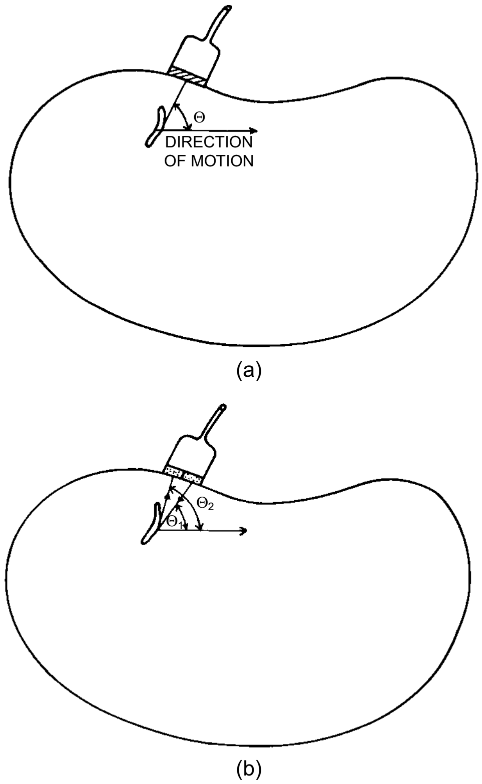
If the source and detector are at the same location and ultrasound is reflected from an object moving toward the location with velocity v, the object acts first as a moving detector as it receives the ultrasound signal and then as a moving source as it reflects the signal. As a result, the ultrasound signal received by the detector exhibits a frequency shift (when $c >> v$)

$$\Delta\nu = 2\nu_0\frac{v}{c}$$

Similarly, for an object moving away from the source and detector, the shift in frequency $\Delta\nu$ is

$$\Delta\nu = 2\nu_0\left(\frac{-v}{c}\right)$$

where the negative sign indicates that the frequency of the detected ultrasound is lower than that emitted by the source.



MARGIN FIGURE 22-2
A: Angle θ between an incident ultrasound beam and the direction of motion of the object. **B:** For a dual probe with separate transmitting and receiving transducers, the angle θ is the average of the angles θ_1 and θ_2 that the transmitted and detected signals make with the direction of motion.

Austrian physicist Christian Doppler (1803–1858) first described the effect that bears his name in a paper published in the proceedings of the Royal Bohemian Society of Learning in 1843. The paper's title translates into English as: "Concerning the colored light of double stars." He explained the shift in color (frequency of light) of stars as their velocities changed relative to earth. While this effect is too small to result in a detectable color shift in measurements available at that time, he demonstrated the reality of this effect in 1845 using sound waves. Two trumpeters who could maintain a steady note with perfect pitch were recruited. One performed at the side of a railway, and the other performed on a moving train. The difference in pitch (frequency of sound) was apparent to listeners.

The discussion above has assumed that the ultrasound beam is parallel to the motion of the object. For the more general case where the ultrasound beam strikes a moving object at an angle θ (Margin Figure 22-2A), the shift in frequency $\Delta\nu$ is

$$\Delta\nu = 2\nu_0\left(\frac{v}{c}\right)\cos\theta \qquad \textbf{(22-1)}$$

A negative sign in this expression would imply that the object is moving away from the ultrasound source and detector and that the frequency of detected ultrasound is shifted to a lower value. The angle θ is the angle between the ultrasound beam and the direction of motion of the object. If the ultrasound source and detector are displaced slightly (Margin Figure 22-2B), then θ is the average of the angles that the transmitted and reflected beams make with the motion of the object. For the small displacement between ultrasound transmitter and detector commonly used in Doppler ultrasound units, this assumption contributes an error of only 2% to 3% in the estimated frequency shift.[3]

Example 22-1

A 10-MHz ultrasound is used for the detection of blood flow. For blood flowing at 15 cm/sec toward the ultrasound source, find the Doppler-shift frequency. Assume that the angle θ is zero so that $\cos\theta = 1$.

$$\begin{aligned}\Delta\nu &= 2\nu_0\left(\frac{v}{c}\right)\cos\theta \\ &= (2)(10\text{ MHz})\left(\frac{15\text{ cm/ sec}}{154{,}000\text{ cm/sec}}\right) \\ &= 1.9\text{ kHz} \qquad \textbf{(22-1)}\end{aligned}$$

The audible frequency range is approximately 20 Hz to 20 kHz. Mechanical vibrations within this frequency range, if delivered with sufficient amplitude, are registered by the ear as audible sound. For physiologic flow rates the Doppler shift (the difference in frequency between sent and received signals) falls in this range and can be used to drive an audio speaker. The listener is not able to derive quantitative velocity information from the audible output but can appreciate changes in signal frequency and intensity. These changes may be caused by variations in the spatial location or angle of the probe with respect to the patient or by variations in the Doppler shift caused by temporal changes in flow within the patient during the cardiac cycle. As flow increases, the frequency of the sound (its "pitch") increases. The listener does not hear a pure single-frequency note but instead hears a complex assortment of frequencies similar to the sound of ocean waves breaking on the seashore. The assortment of frequencies, referred to as the spectral bandwidth of the Doppler signal, is discussed in detail later in this chapter.

The term involving $\cos\theta$ in the Doppler-shift equation [Eq. (22-1)] describes the variation in frequency shift as the sonation angle (the angle between the Doppler probe and the moving material) is varied. The shift is greatest when the longitudinal axis of the probe is oriented along the direction of motion. This occurs when the transmitted beam of sound is parallel (sonation angle = 0 degrees) or opposed to (sonation angle = 180 degrees) the direction of motion. Theoretically, no Doppler shift occurs when the probe is exactly perpendicular to the direction of motion. In practice, a small Doppler shift may be detected when the probe appears to be perpendicular to the direction of blood flow in a vessel. The small shift is due to several phenomena such as transverse components of flow in the vessel, detection of signals from side lobes, and so on.

The presence of the cosine term in the Doppler-shift equation complicates the direct measurement of velocity by the Doppler method. The only quantity that is measured directly by a typical Doppler ultrasound unit is Doppler shift. Determination of the velocity of blood in units such as centimeters per second requires an estimate of the sonation angle. Errors in the estimate of this angle (using on-screen cursors) limit the accuracy of velocity estimates. This error is more significant at larger angles (near 90 degrees) than at smaller angles.

An error in a sonographer's ability to estimate the Doppler angle in a real-time B-mode display affects the estimated value of blood velocity. The same error has a greater effect at angles near 90° than at angles near 0°. The error occurs because calculation of velocity from the Doppler frequency shift [Eq. (22-1)] involves multiplication by the cosine of the Doppler angle. This affect is appreciable even if the error is small.

If we assume an error of ± 3 degrees, then at an actual Doppler angle of 10° the sonographer's estimate will lead to a variation in the value of the cosine function from cos (7°) = 0.993 to cos (13°) = 0.974 where the true value of cos (10°) = 0.985. The error is less than ± 1%.

At an actual Doppler angle of 80° the same error of 3° leads to a variation in the value of the cosine function from cos (77°) = 0.225 to cos (83°) = 0.122 where the true value of cos (80°) = 0.174. Here, the error is approximately 30%!

Example 22-2

A 10-MHz ultrasound probe detects a 1.4-kHz Doppler shift at a sonation angle estimated as 45 degrees. Find the estimated velocity of blood flow and the percent error in the estimate if the angle of sonation is incorrect by as much as 3 degrees:

$$\Delta\nu = \frac{2\nu_0 \, v \cos\theta}{c}$$

$$v = \frac{\Delta\nu c}{2\nu_0 \cos\theta}$$

$$= \frac{(1.4 \times 10^3 \text{ sec}^{-1})(1.54 \times 10^5 \text{ cm/sec})}{2(10 \times 10^6 \text{ sec}^{-1})} \frac{1}{\cos\theta}$$

$$= (10.8 \text{ cm/sec}) \frac{1}{\cos\theta}$$

If $\theta = 45$ degrees

$$v = \frac{10.8}{0.707} = 15.3 \text{ cm/sec}$$

But if θ actually equaled 48 degrees instead of 45, then

$$v = \frac{10.8}{0.669} = 16.1 \text{ cm/sec}$$

The percent error is

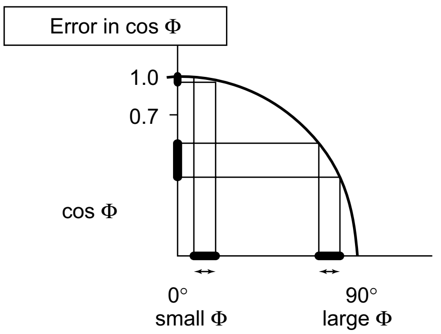
$$\% \text{ error} = \frac{16.1 - 15.3}{15.3}(100) = 5.5\%$$

The error is slightly smaller if θ were actually 42 degrees.

Determining the volumetric flow of blood in units of cubic centimeters per second requires an estimate of the area of the vessel as well as the sonation angle. Furthermore, it assumes measurement of average quantities that are not always valid. Volumetric flow Q is the product of the average velocity v and the cross-sectional area A of the vessel:

$$Q = vA \tag{22-2}$$

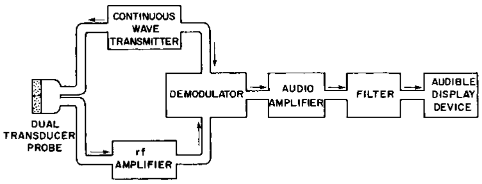
In clinical measurements of volumetric flow, several sources of error reduce the accuracy below what would be expected based upon the intrinsic precision of Doppler-shift measurements. The measurement of vessel area is typically based upon measurement of vessel diameter in the image. However, the angle at which the image plane "cuts through" the vessel may cause either underestimation or overestimation of diameter. Also, the Doppler unit may not sample the average velocity in a vessel. Flow in a vessel (as in a mountain stream) does not have a uniform cross-section. It is usually greatest at the center and decreases to near zero at the vessel wall. Complex flow "profiles" are possible, particularly if a stenosis, bifurcation, or plaque formation is present. An accurate estimate of average velocity requires sampling of velocity at different radii within the vessel (at several points across the "stream"). It is possible to make some simplifying assumptions. For example, in laminar (nonturbulent) flow the average velocity is half the maximum value.[4]



MARGIN FIGURE 22-3
Because of the dependence of the Doppler shift upon the cosine of the Doppler angle, calculation of the velocity of an object from an observation of its Doppler shift requires estimation of the Doppler angle. This estimate (usually acquired through observation of a freeze frame of a real-time B mode ultrasound image) could be off by a few degrees. In the graph shown here we see that if the few-degree error (horizontal axis) occurs at a small Doppler angle (close to parallel to the direction of flow), the error introduced by the cosine function (vertical axis) is relatively small. If the same few degree error occurs at a larger Doppler angle (close to perpendicular to the direction of flow), then the error introduced by the cosine function is greater.

Example 22-3

Find the volumetric flow in a 3-mm-diameter vessel in which the average velocity (v) of blood is estimated to be 10 cm/sec.



MARGIN FIGURE 22-4
Diagram of a continuous-wave Doppler unit.

First, find the cross-sectional area (A) of the vessel.

$$A = \pi\left(\frac{d}{2}\right)^2 = 3.14\left(\frac{0.3 \text{ cm}}{2}\right)^2$$
$$= 0.07 \text{ cm}^2$$

The volumetric flow (Q) is

$$\begin{aligned} Q &= vA \\ &= (10 \text{ cm/sec})(0.07 \text{ cm}^2) \\ &= 0.7 \text{ cm}^3\text{/sec} \end{aligned} \tag{22-2}$$

A continuous-wave Doppler unit is diagramed in Margin Figure 22-4. The Doppler probe contains separate transmitting and receiving transducers. Application of a continuous electrical signal to the transmitting transducer produces a continuous ultrasound beam at a prescribed frequency. As the beam enters the patient, it is reflected from stationary and moving surfaces, and the reflected signals are returned to the receiver. Suppose that two signals are received, one directly from the transmitting transducer and the other from a moving surface. These signals interfere with each other to produce a net signal with a "beat frequency" equal to the Doppler-shift frequency. This Doppler-shift signal is amplified in the radio-frequency (RF) amplifier. The amplified signal is transmitted to the demodulator, where most of the signal is removed except the low-frequency beat signal. The beat signal is amplified in the audio amplifier and transmitted through the filter (where frequencies below the beat frequency are removed) to the loudspeaker or earphones for audible display. In practice, many moving interfaces reflect signals to the receiver, and many beat frequencies are produced.

Pulsed Doppler

Accurate estimates of the location of the source of a Doppler shift in a patient are difficult to achieve with continuous-wave Doppler. The source is obviously somewhere along the extended axis of the ultrasound transducer (direction of sonation), but exactly where is uncertain. To localize the source of the Doppler shift, a method known as range gating may be used. In this method the sending transducer is shut off, and a preset interval of time, the first range gate, is allowed to elapse before the return signal is processed to determine the frequency shift. The Doppler shift is then analyzed until a second range gate is encountered. With the assumption of an average speed of sound, the range gate settings define a specific Doppler sampling region over a particular range of depth. The time between the two gates determines the extent of the sampling region, and variation of the elapsed time before the first gate determines the sampling depth. With a speed of sound for soft tissue of 1540 m/sec, a range factor of 13 μs of elapsed time corresponds to 1 cm of depth. This range factor takes into account the "round trip" of 2 cm that sound travels (i.e., from the transducer to the reflector and back again) for each centimeter of depth.

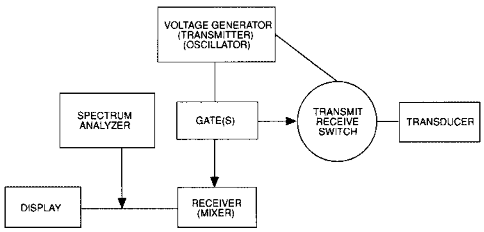
Example 22-4

Range gates are set at 39 and 45 μsec with respect to a transmitted pulse of ultrasound. Determine the dimensions of the Doppler sampling region within soft tissue.

$$\frac{39\ \mu\text{sec (1st gate)}}{13\ \mu\text{sec/cm}} = 3 \text{ cm from the transducer}$$

$$\frac{45\ \mu\text{sec (2nd gate)}}{13\ \mu\text{sec/cm}} = 3.5 \text{ cm from the transducer}$$

The Doppler sampling region begins at a depth of 3 cm and extends to a depth of 3.5 cm. We could state that there is a 0.5-cm sampling region centered at a depth of 3.25 cm. The width of the sampling region is roughly equal to the ultrasound beam width at that depth.



MARGIN FIGURE 22-5
Diagram of a pulsed Doppler unit.

In Doppler systems that are used in conjunction with imaging, the position of the sampling volume is usually indicated by cursors positioned on the display. The ultrasound computer performs the electronic equivalent of the calculations in Example 22-4 to determine where the cursors should appear on the image.

Range gating requires that the ultrasound be started and stopped (i.e., pulsed) during the signal acquisition process so that the time to echo may be determined unambiguously. The use of pulsed Doppler not only allows range gating but also permits imaging pulses to be interspersed with Doppler pulses. Thus a real-time B-mode image of the patient may be obtained from every other pulse. In continuous-wave Doppler there is, by definition, no time in which to send imaging pulses, because there is no gap in the Doppler process.

A simplified diagram of a pulsed Doppler ultrasound system is shown in Margin Figure 22-5. The main differences between the pulsed Doppler system in this illustration and the continuous-wave system of Margin Figure 22-4 are that the pulsed system includes range gates and a transmit–receive switch similar to that in B-mode systems. Units that intersperse Doppler and imaging pulses incorporate components found in standard B-mode systems with both Doppler and B-mode components connected to the voltage generator or oscillator so that the sending of Doppler and imaging pulses can be coordinated.

There are techniques for determining velocity in ultrasound images that do not use the Doppler frequency shift effect. In "time domain" analysis, the return signals along scan lines are compared in a process known as "autocorrelation analysis" to determine if a signal anywhere in the image in one frame appears at a different location in another frame. This information is then color-coded using any of the techniques that are used in color Doppler.

Spectral Analysis

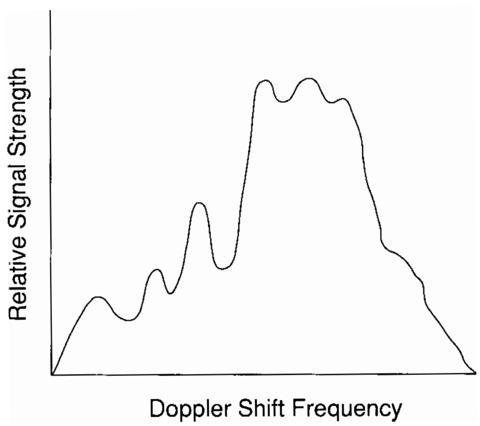
The Doppler-shift signal contains a wealth of information. One of the fundamental challenges of Doppler ultrasound is to display that information in a useful fashion. Doppler shift in a blood vessel is produced by the flow of red blood cells (RBCs). However, within a single vessel there is an assortment (distribution) of RBC velocities. Also, the velocity of any given cell varies with time according to the cardiac cycle. Three methods for visual display of information are discussed here: frequency spectrum, spectral trace, and color mapping.

Frequency Spectrum

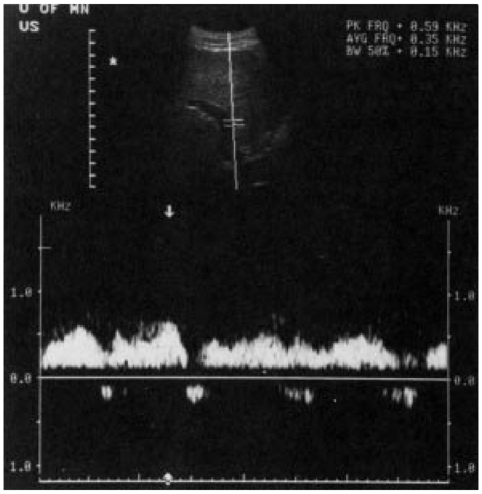
A frequency spectrum (Margin Figure 22-6) is a histogram in which the horizontal axis depicts Doppler-shift frequency and the vertical axis shows the relative contribution of each frequency to the observed Doppler shift. For simplicity, the frequency spectrum can be thought of as a histogram of the relative distribution of RBC velocities in a blood vessel. The peak of the spectrum shows the Doppler shift associated with the greatest number of RBCs. The frequency spectrum is a "snapshot" depicting the conditions of flow in the sampling volume at a particular moment in time. To determine temporal variations, a display such as the flow plot is required.

Spectral Trace

A spectral trace (Margin Figure 22-7) is a graph of frequency versus time that depicts the manner in which the value of Doppler shift varies with time. The value indicted may be the average, minimum, or maximum Doppler shift, and a distribution of values may be indicated by a vertical bar. The spectral trace may be displayed as a Doppler-shift frequency trace or as a velocity trace. The velocity trace is obtained by using the center frequency of the probe and the sonation angle measured by the operator to convert the Doppler-shift frequency scale a velocity scale (see Example 22-2). Audible sound may be broadcast by speakers as the trace is displayed in real time



MARGIN FIGURE 22-6
Doppler-shift frequency spectrum showing the relative contribution of frequencies to the Doppler shift during a brief interval of time.



MARGIN FIGURE 22-7
Doppler spectral trace showing Doppler shift as a function of time. The distribution of Doppler shifts at any given time is the vertical "thickness" of the trace. Thus, the Doppler spectral trace shows changes in the frequency spectrum over time.

by using a "moving pen" or "strip chart" type of display. Cardiac information such as electrocardiographic (ECG) tracings may also be displayed to allow the viewer to coordinate features in the spectral trace with timing of the cardiac cycle. Some systems also allow the user to display a frequency spectrum for a selected moment of time during display of the spectral trace.

Color Mapping

Both frequency spectra and spectral traces may be displayed along with B-mode images by dividing the display screen into separate "windows" (compartments). Cursors on the image indicate the Doppler sampling volume. It is possible to indicate flow directly within the image by color-coding the pixels. In color mapping or color-flow Doppler, pixels in the B-mode image that correspond to regions in the patient where flow is present are assigned hues (colors such as red, blue, green) according to an arbitrary color scale. For example, red may indicate a large positive Doppler shift, orange a smaller positive shift, yellow a small negative shift, and so on.

Alternatively, the color scale may involve only two hues. One of the most common two-hue systems uses red for a positive Doppler shift and blue for a negative Doppler shift. The magnitude of the Doppler shift is encoded by the "saturation" of the color. Saturation is the degree to which the hue is present. A totally unsaturated color is white. It emits a broad distribution of all frequencies of visible light. Higher saturation implies that more of the frequencies in the spectrum are of a single frequency, interpreted by our retina as a purer color. The saturation encoding in color Doppler is usually arranged such that a greater shift (higher velocity either toward or away from the transducer) is encoded as a less saturated hue. For example, a small positive Doppler shift would be indicated by a dark red, whereas a larger positive Doppler shift would be indicated by a lighter, less saturated red.

In color Doppler, the most commonly used color scale codes flow away from the transducer as blue and flow toward the transducer as red. When such a "blue away red toward" (BART) display is used, it is sometimes necessary to remind observers that the colors are not necessarily correlated with the oxygenation (venous or arterial) status of the blood.

Whatever color scale is used in color Doppler, it is desirable to avoid color-coding the very slow velocities of tissues that occur in the patient as pliable tissues are pushed by the transducer and as the transducer itself is moved. Detection of such motions is not usually the purpose of the exam. Thus, most color scales do not assign color to very small Doppler shifts and display a black band at the center of the color scale indicator. The width of the black band may be changed by the operator to vary the range of velocities that will be color-coded.

The large variety of color scales that could be used in color Doppler, as well as the variation in perception of color from one individual to another, is an intriguing problem for optimization of visual perception.

Quantitative Use of Doppler

Various attempts have been made to correlate information obtained from Doppler displays, particularly spectral traces, with pathology. Simple measures such as the peak systolic or end-diastolic average velocities are probably too crude to differentiate normal and disease states, in part because of the wide range of normal values in a population and also because of the difficulties inherent in obtaining quantitative values for velocity. It is possible to construct ratios of features of spectral traces so that conversion factors such as the angle of sonation are the same for both features and therefore cancel out. With the symbol A used for peak systolic and B for end-diastolic frequency shift, some indices that have been used with varying degrees of success include A/B,[5] $(A - B)$/mean,[6] and $A - B/A$ (known as Pourcelot's or the resistivity index).[7] Still, attempts to provide definitive diagnoses on the basis of indices alone have not been generally successful. Such methods appear more useful for longitudinal studies than for specific diagnoses.[8]

LIMITATIONS OF DOPPLER SYSTEMS

Pulse Repetition Frequency

The use of pulsed Doppler limits the maximum velocity that can be detected. Each pulse reflected from the RBCs is a sample of the frequency shift. The greater the rate of sampling or pulsing, the better the measurement of the shift. Information theory suggests that an unknown periodic signal must be sampled at least twice per cycle to determine even rudimentary information such as the fundamental frequency.[4] Thus the rate of pulsing [the pulse repetition frequency (PRF)] of pulsed Doppler must be at least twice the Doppler-shift frequency produced by flow. If not, the frequency shift is said to be "undersampled"—that is, sampled so infrequently that the frequency reported by the instrument is erroneously low, an artifact known as "aliasing."

Example 22-5

A stenosis produces a high-speed jet in a vessel. The maximum velocity of RBCs in the stenosis is 80 cm/sec. Find the minimum PRF that must be used to avoid aliasing for pulsed Doppler at 7.5 MHz. Assume that the angle of sonation is 0 degrees.

The Doppler shift is

$$\begin{aligned}\Delta\nu &= \frac{2\nu_0\, v\cos\theta}{c}\\ &= \frac{2(7.5\times 10^6\ \text{sec}^{-1})(80\ \text{cm/sec})(1)}{1.54\times 10^5\ \text{cm/sec}}\\ &= 7.8\ \text{kHz}\end{aligned} \quad \textbf{(22-1)}$$

The minimum PRF required to prevent aliasing is twice the Doppler-shift frequency, or 15.6 kHz.

In color Doppler, aliasing is characterized by colors from one end of the color spectrum appearing adjacent to colors from the opposite end. For example, an abrupt transition from blue to red with no dark band between the two colors signals aliasing. A dark band would indicate momentary cessation of flow which would have to occur if there was a change in flow direction.

There is a practical limit to the PRF of an ultrasound system. To obtain information at a given depth in the patient, the system must "wait and listen" over the time required for sound to travel from the transducer to that depth and back again. If an echo returns after another pulse has been sent, it is difficult to determine which travel time to use in determining the range. Because the travel time required for sound waves in soft tissue is 13 μsec/cm of depth, the maximum PRF for a given total depth of imaging [field of view (FOV)] can be computed.

Example 22-6

Find the field of view (FOV) if the PRF of 15.6 kHz calculated from Example 22-5 is used for imaging.

If the PRF is 15.6 kHz, then the wait-and-listen time between pulses, the pulse repetition period (PRP), is its reciprocal.

$$\begin{aligned}\text{PRP} = \frac{1}{\text{PRF}} = \frac{1}{15.6\times 10^3\ \text{sec}^{-1}} &= 6.41\times 10^{-5}\ \text{sec}\\ &= 64.1\ \mu\text{sec}\end{aligned}$$

The field of view (FOV) is then

$$\text{FOV} = \frac{64.1\ \mu\text{sec}}{13\ \mu\text{sec/cm}} = 4.9\ \text{cm}$$

Clearly, the PRF required to measure the maximum flow in a high-speed jet is not routinely available in imaging systems because it would severely limit the field of view.

In echocardiography, the appearance of very rapidly moving blood that spurts from heart valves that do not close properly (known as "jets") may be enhanced in color flow Doppler. One technique that enhances jets is to color code signals with an extremely high Doppler shift in a color that is markedly different from the traditional red and blue (e.g., green or yellow). Then, when a jet appears in a real time image, a spurt of bright color is immediately apparent.

To avoid aliasing, most pulsed Doppler systems have a "high-PRF" option that permits operation at PRFs that may exceed those practical for imaging. An alternative is to switch to continuous-wave Doppler because it is not limited in the same way that pulsed Doppler is limited by the sampling process. In some cases, it may be possible to switch to a lower-frequency transducer so that the Doppler shift is lower. Several techniques to circumvent the PRF requirement have been proposed but are still in the research stages. These include pulse-coding schemes and the use of nonperiodic pulses.[9,10]

Spectral Broadening

There are many barriers to interpreting Doppler shift as a measure of a fundamental parameter such as velocity. Just as other imaging modalities are limited by noise, Doppler ultrasound is also limited by extraneous information or noise. This extraneous information may cause a "spectral broadening" of the range of Doppler-shift frequencies displayed by the unit. Spectral broadening compromises the interpretation of the width of the frequency spectrum as an indication of the range of RBC velocities. Spectral broadening can also be caused by other factors that may be lumped together as deterministic errors (see Chapter 11). These include the bandwidth or range of frequencies in the transmitted signal, the presence of adjacent vessels or flow within the fringes of the sampling volume, and tissue and wall motion (this low-velocity, low-frequency component is usually eliminated by electronic filtering known as "wall filtering").

Nondeterministic or random spectral broadening may be caused by electronic noise resulting from random fluctuations in electronic signals.

PROBLEMS

*22-1. Estimate the frequency shift for a 10-MHz ultrasound source moving toward a stationary detector in water ($c = 1540$ m/sec) at a speed of 5 cm/sec. Is the frequency shifted to a higher or lower value?

*22-2. Estimate the frequency shift for an object moving at a speed of 10 cm/sec away from a 10-MHz source and detector occupying the same location.

22-3. What is meant by a beat frequency?

22-4. Identify some of the major medical applications of Doppler systems.

*22-5. Find the minimum pulse repetition frequency required to prevent aliasing when measuring a velocity of 5 cm/sec with pulsed Doppler ultrasound at 5 MHz for a sonation angle of zero degrees. How would this PRF change if:
a. The angle of sonation was increased?
b. The center frequency was increased?

*For those problems marked with an asterisk, answers are provided on p. 493.

SUMMARY

- The Doppler equation is $\Delta\nu = 2\nu_0 \frac{v}{c} \cos\theta$.
- As the Doppler angle increases from zero to 90 degrees, the Doppler shift-decreases and errors in estimation of the angle cause greater inaccuracies in velocity estimation.
- Volumetric flow Q, average velocity v, and area A are related as follows:

$$Q = vA$$

- The main components of a Doppler ultrasound imager include
  - Transducer
  - Transmit/receive switch
  - Voltage generator (transmitter or oscillator)
  - Gate system

  - Receiver (mixer)
  - Spectrum analyzer
  - Display
- A Doppler spectral trace is a plot that shows the variation in time of a brightness modulated Doppler spectra.
- In color Doppler, the magnitude of frequency shift is indicated by saturation and the specific colors indicate direction of flow.
- Spectral broadening may indicate that a range of velocities exists in the sample volume. It may also be caused by
  - Bandwidth of the transmitted pulse
  - Signal from adjacent vessels
  - Tissue and vessel wall motion
  - Electronic noise

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