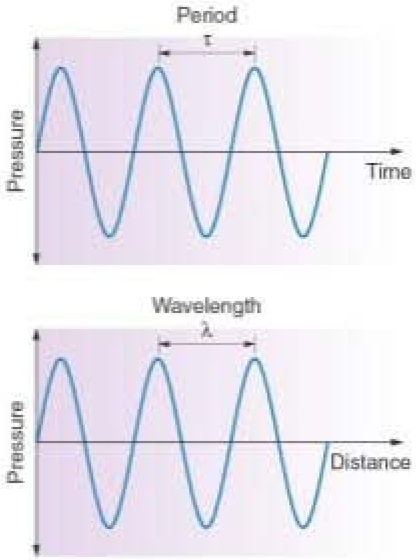

FIGURE 1-1 Sinusoidal wave depicted on the time axis and distance axis. The time to complete one cycle is the period (τ). The distance to complete one cycle is the wavelength (λ).

The acoustic velocity (c) of a medium can be determined once the density (ρ) and the compressibility (K), or bulk modulus (β), are known. Equation 1.3 demonstrates the relationship of the three physical properties.

Density, compressibility, and bulk modulus are not independent of one another. Typically, as density increases, compressibility decreases and bulk modulus increases. However, compressibility and bulk modulus typically vary more rapidly than does density, and they dominate in Equation 1.3.

$$c=\frac{1}{\sqrt{K\rho}}=\frac{\sqrt{\beta}}{\sqrt{\rho}} \tag{1.3}$$

The acoustic velocity in different media can be determined by applying the equations to practice. For example, water at 30°C has a density of 996 kg/m^3 and a bulk modulus of 2.27 × 10^9 N/m^2.[2] Inserting these values into Equation 1.3 yields an acoustic velocity of 1509 m/s in water. Values for density and bulk modulus have been characterized extensively and can be found in the literature.[2] A summary of relevant tissue properties is given in Table 1-1. The acoustic velocity is not dependent on the frequency of the propagating wave (i.e., acoustic waves of different frequencies all propagate with the same acoustic velocity within the same medium).[3]

Ultrasound Interactions in Tissue

Ultrasound imaging of tissue is achieved by transmitting short pulses of ultrasound energy into tissue and receiving reflected signals. The reflected signals that return to the transducer represent the interactions of a propagating ultrasound wave with tissue. A propagating ultrasound wave can interact with tissue, and the results are *reflection*, *refraction*, *scattering*, and *absorption*.

Reflection

Specular reflections of ultrasound occur at relative large interfaces (greater than one wavelength) between two media of

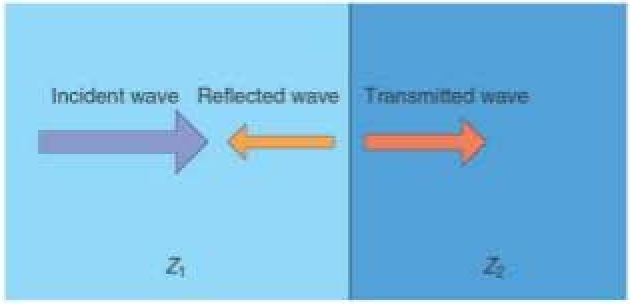

FIGURE 1-2 Reflection of an ultrasound wave at normal incidence to an interface between two media with different acoustic impedances (Z).

differing acoustical impedances. At this point, it is important to introduce the concept of *acoustic impedance*. The acoustic impedance (Z) of a medium represents the resistance to sound propagating through the medium and is the product of the *density* (ρ) and the *velocity* (c):

$$Z=\rho c \tag{1.4}$$

Sound will continue to propagate through a medium until an interface is reached where the acoustic impedance of the medium in which the sound is propagating differs from the medium that it encounters. At an interface where an acoustic impedance difference is encountered, a proportion of the ultrasound wave will be reflected back toward the transducer, and the rest will be transmitted into the second medium. The simplest case of reflection and transmission occurs when the propagating ultrasound wave is perpendicular (90 degrees) to the interface (Figure 1-2). In this case, the percentage of the incident beam that is reflected is as follows:

$$\%\text{reflected}=\left(\frac{Z_2-Z_1}{Z_2+Z_1}\right)^2\times 100 \tag{1.5}$$

The percentage of the incident beam that is transmitted is as follows:

$$\%\text{transmitted}=100-\%\text{reflected} \tag{1.6}$$

Refraction

When the incident beam arrives at the interface at an angle other than 90 degrees, the transmitted beam path diverges from the incident beam path because of refraction

TABLE 1-1

PHYSICAL PROPERTIES OF TISSUE

Tissue or Fluid	Density (kg/m^3)	Bulk Modulus (×10^9 N/m^2)	Acoustic Velocity (m/s)
Water (30°C)	996	2.27	1509
Blood	1050-1075	2.65	1590
Pancreas (pig)	1040-1050	2.63	1591
Liver	1050-1070	2.62	1578
Bone, cortical	1063-2017	28.13	3760

Adapted from Duck FA. Physical Properties of Tissue. *London: Academic Press; 1990.*

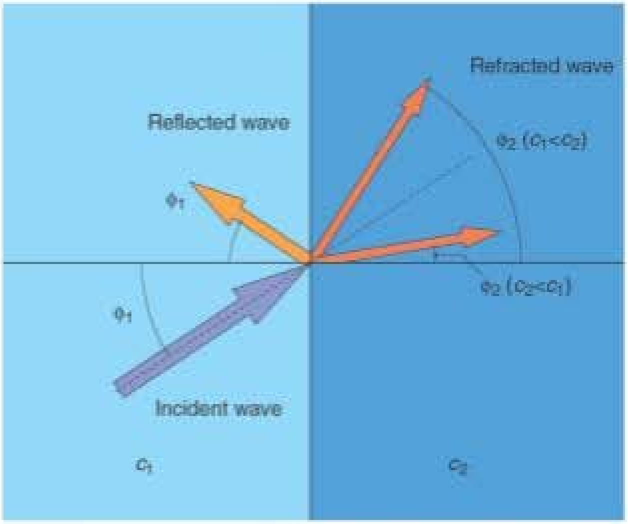

FIGURE 1-3 Refraction and reflection of an incident wave that is not normal to the interface between media with different acoustic velocities (c). The angle of reflection is identical to the angle of incidence. The angle of the refracted wave is dependent on the acoustic velocities of the two media and can be determined by applying Snell's law (see text).

(Figure 1-3). The angle at which the transmitted beam propagates is determined by Snell's law:

$$\frac{\sin\varphi_1}{\sin\varphi_2}=\frac{c_1}{c_2} \tag{1.7}$$

The angle of *refraction* is determined by the *acoustic velocities* in the incident (c_1) and transmitted (c_2) media. There are three possible scenarios for a refracted beam, depending on the relative speeds of sound between the two media: (1) if $c_1 > c_2$, the angle of refraction will be bent toward normal ($\phi_1 > \phi_2$); (2) if $c_1 = c_2$, the angle of refraction will be identical to the angle of incidence, and the beam will continue to propagate without diverging from its path; (3) if $c_1 < c_2$, the angle of refraction will be bent away from normal ($\phi_1 < \phi_2$). Refraction of the ultrasound beam can lead to imaging artifacts that are discussed later in the chapter.

Scattering

Scattering, also termed *nonspecular reflection*, occurs when a propagating ultrasound wave interacts with different components in tissue that are smaller than the wavelength and have different impedance values than the propagating medium.[4] Examples of scatterers in tissue include individual cells, fat globules, and collagen. When an ultrasound wave interacts with a scatterer, only a small portion of the acoustic intensity that reflects off of the scatterer is reflected back to the transducer (Figure 1-4). In addition, a signal that has undergone scattering by a single scatterer will usually undergo multiple scattering events before returning to the transducer. Scattering occurs in heterogeneous media, such as tissue, and is responsible for the different echotextures of organs such as the liver, pancreas, and spleen. Tissue containing fat or collagen scatters ultrasound to a greater degree than do other tissues, and this is why lipomas and the submucosal layer of the gastrointestinal tract appear hyperechoic (bright) on ultrasound imaging.[4]

Multiple reflections from nonspecular reflectors within the tissue returning to the transducer result in a characteristic acoustic speckle pattern, or echotexture, for that tissue.[4] Because speckle originates from multiple reflections and does not represent the actual location of a structure, moving the transducer will change the location of the speckle echoes while maintaining a similar speckle pattern. In addition, the noise resulting from acoustic speckle increases with increasing depth as a result of the greater number of signals that have undergone multiple reflections from nonspecular reflectors returning to the transducer.

Absorption

Ultrasound energy that propagates through a medium can be absorbed, resulting in the generation of heat. The *absorption* of ultrasound energy depends on tissue properties and is highly frequency dependent. Higher frequencies cause more tissue vibration and result in greater absorption of the ultrasound energy and more heat generation.

Ultrasound Intensity

The *intensity* of the ultrasound signal is a parameter that describes the power of the ultrasound signal over a cross-sectional area. As ultrasound waves propagate through tissue, the intensity of the wave becomes attenuated. Attenuation is the result of effects of both scattering and absorption of the ultrasound wave.[1] The *attenuation coefficient* (a) is a function of frequency that can be determined experimentally, and it increases with increasing frequency. The frequency of the ultrasound pulse affects both the depth of penetration of the

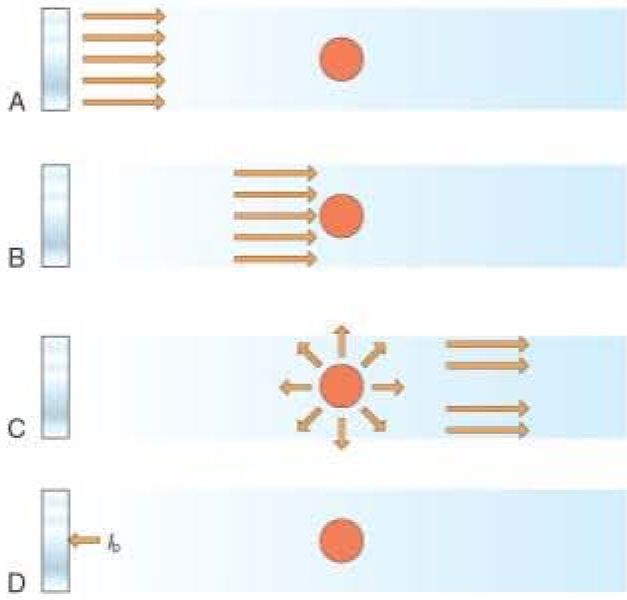

FIGURE 1-4 Schematic representation of single scattering. Scattering occurs from an interface that is smaller than the wavelength of the propagating ultrasound signal. The transducer is responsible for sending and receiving the signal. I_b is the back-scattered intensity that will propagate back to the transducer. **A**, The ultrasound signal is transmitted by the transducer and propagates toward the scatterer. **B**, The pulse reaches the scatterer. **C**, The incident acoustic intensity is scattered in different directions. **D**, The back-scattered energy received by the transducer is only a small fraction of the incident acoustic intensity that is scattered.

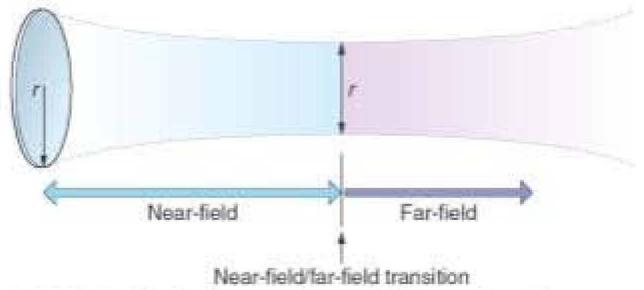

FIGURE 1-7 Single-element unfocused disk transducer. In a nonattenuating medium, an unfocused transducer has a self-focusing effect with the diameter of the ultrasound beam at the focus equal to the radius of the transducer (r). The location of the beam waist occurs at the near-field/far-field transition.

wavelength of ultrasound in the propagating medium. Equation 1.9 demonstrates that, as the radius of the transducer decreases, the focal length is reduced if the frequency remains constant. In addition, for a constant radius, increasing the wavelength (i.e., decreasing the frequency) also reduces the focal length. However, in attenuating media such as tissue, this self-focusing effect is not seen, and the beam width in the near-field is approximately equal to the diameter of the transducer (Figure 1-8). The beam width then rapidly diverges in the far-field.

Focusing. A single-element transducer can be focused by fabricating the transducer with a concave curvature (spherically curved) or by placing a lens over a flat disk transducer. Focusing is used to improve the lateral resolution and results in a narrow beam width at the focal length (distance from the transducer to the location of the beam width that is most narrow). However, the degree of focusing affects the depth of focus (the range where the image is in focus) and the focal length. For weak focusing, the focal length is long, as is the depth of focus. Conversely, for a beam that is highly focused, the focal length is short, as is the depth of focus (Figure 1-9).

Arrays. Multiple single-element transducers can be combined in several different configurations. The linear array configuration is the most widely employed clinically. The array is composed of multiple identical crystals that are controlled electronically (Figure 1-10). They can be fired individually in sequence or in groups, depending on the imaging algorithm.

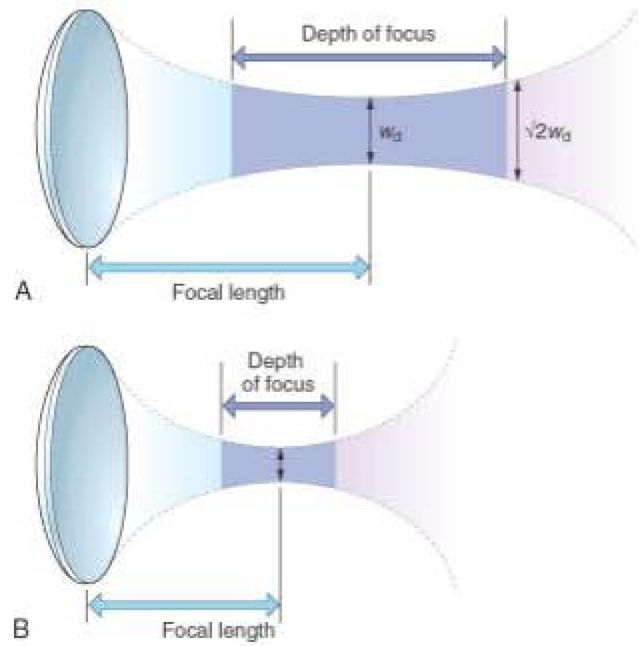

FIGURE 1-9 Effect of focusing. Focusing increases lateral resolution by decreasing the beam waist in the focal region *(highlighted in blue)*. The depth of focus is the distance between where the diameter of the beam is equal to $\sqrt{2}w_0$, where w_0 is the diameter of the beam at the waist or focus. The degree of focusing influences the focal length, as well as the depth of focus. This figure compares two transducers of equal diameters with different degrees of focusing. The transducer in (**A**) exhibits weak focusing, whereas that in (**B**) exhibits strong focusing. The diameter of the waist at the focus is narrower with strong focusing, and this leads to improved lateral resolution in the focal region. However, the trade-off is a decrease in the depth of focus with rapid divergence of the beam beyond the focus. In addition, the focal length is much shorter (i.e., the focus is closer to the transducer) for the highly focused transducer.

This configuration allows for electronic focusing at different depths based on the timing of the excitation of the individual transducer crystals.

Processors

Figure 1-5 is a block diagram of the components of an ultrasound imaging system. The main components are the ultrasound transducer, processor, and display. Within the processor

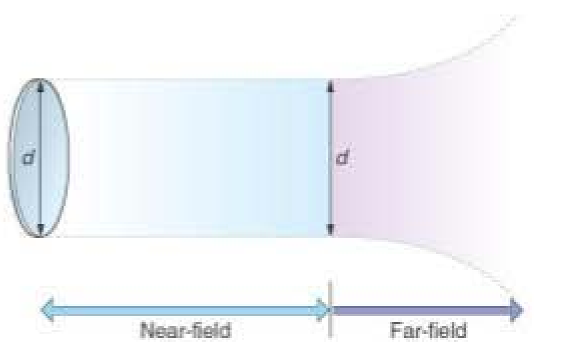

FIGURE 1-8 Single-element unfocused disk transducer. In an attenuating medium, the beam width of an unfocused transducer is approximately equal to the diameter of the transducer (d) until the near-field/far-field transition. The beam then rapidly diverges in the far-field.

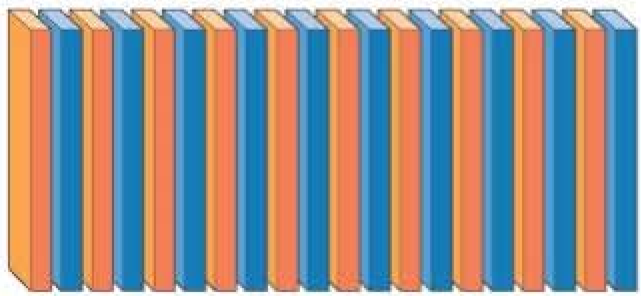

FIGURE 1-10 Configuration of a linear array transducer. This configuration consists of several rectangular elements, which are controlled individually. The sequence and timing of excitation of each individual element dictate the beam pattern that is transmitted from the array.

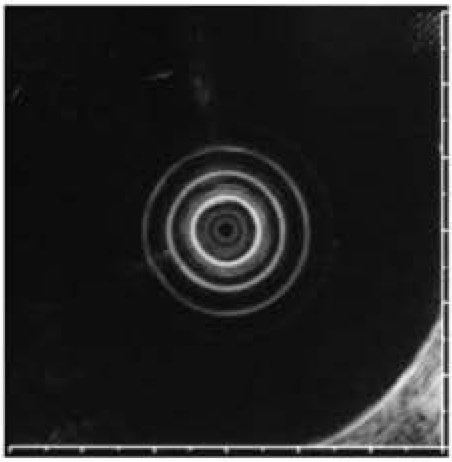

FIGURE 1-16 EUS image of reverberation artifact resulting from multiple reflections from the transducer housing. The concentric rings are equally spaced, with the intensity of the rings decreasing as the distance from the transducer increases.

Acoustic Shadowing

Acoustic shadowing is a form of a reflection artifact that occurs when a large impedance mismatch is encountered. When such a mismatch is encountered, a majority of the transmitted pulse is reflected with minimal transmission. This results in a hyperechoic signal at the interface with no echo signal detected beyond the interface, thus producing a shadow effect. This finding is useful in diagnosing calcifications in the pancreas (Figure 1-20) and gallstones in the gallbladder (Figure 1-21).

Acoustic shadowing can also result from refraction occurring at a boundary between tissues with different acoustic velocities, especially if the boundary is curved (e.g., tumor or cyst). As discussed earlier, refraction of an ultrasound beam occurs when the angle of incidence is not normal to the boundary between tissues with different acoustic velocities, with resulting bending of the ultrasound beam. Because the ultrasound beam is redirected at this boundary, some regions of the tissue are not interrogated by the ultrasound beam, and the result is an acoustic shadow (Figure 1-22).[8]

FIGURE 1-17 EUS image of reverberation artifact *(arrow)* resulting from multiple reflections from an air bubble in the water-filled balloon. The intensity of the artifact does not decrease as rapidly as the reverberation artifact *(arrowhead)* from the transducer housing. This is because the impedance mismatch of the air-water interface is much greater than the transducer housing interface, with resulting reflected signals of greater intensity.

FIGURE 1-18 Reflection or mirror image artifact. A mirror image of the transducer *(arrowhead)* and gastric wall is produced by the reflection of the ultrasound signal from the interface between water and air *(arrow)* within the gastric lumen.

Through Transmission

Through transmission is the enhancement of a structure beyond a fluid-filled structure such as a cyst. The structure beyond a fluid-filled structure demonstrates increased enhancement because the intensity of transmitted ultrasound undergoes less attenuation as it propagates through the cyst and as the reflected signal returns to the transducer. This finding is useful in diagnosing fluid-filled structures such as a cyst or blood vessel (Figure 1-23).

Tangential Scanning

If the thickness of a structure is being measured, it is important that the ultrasound beam is perpendicular to the structure. If the transducer is at an angle other than 90 degrees to the structure, the thickness will be overestimated.[8] This is particularly important when assessing the thickness of the layers of the gastrointestinal (GI) tract wall and in staging tumors of the GI tract. On radial scanning examination of the GI tract, this artifact can be identified because the thicknesses of the wall layers will not be uniform throughout the image (Figure 1-24). When staging tumors involving the GI tract wall, tangential imaging can result in overstaging of the tumor. To avoid this artifact, the endoscope tip should be maneuvered to maintain the proper orientation such that the plane of imaging is normal (at 90 degrees) to the structure being imaged.

Side Lobe Artifacts

Side lobes are off-axis secondary projections of the ultrasound beam (Figure 1-25).[3] The side lobes have reduced intensities compared with the main on-axis projection; however, they can produce image artifacts. Usually, on-axis reflections are greater in intensity than side lobe reflections and thereby

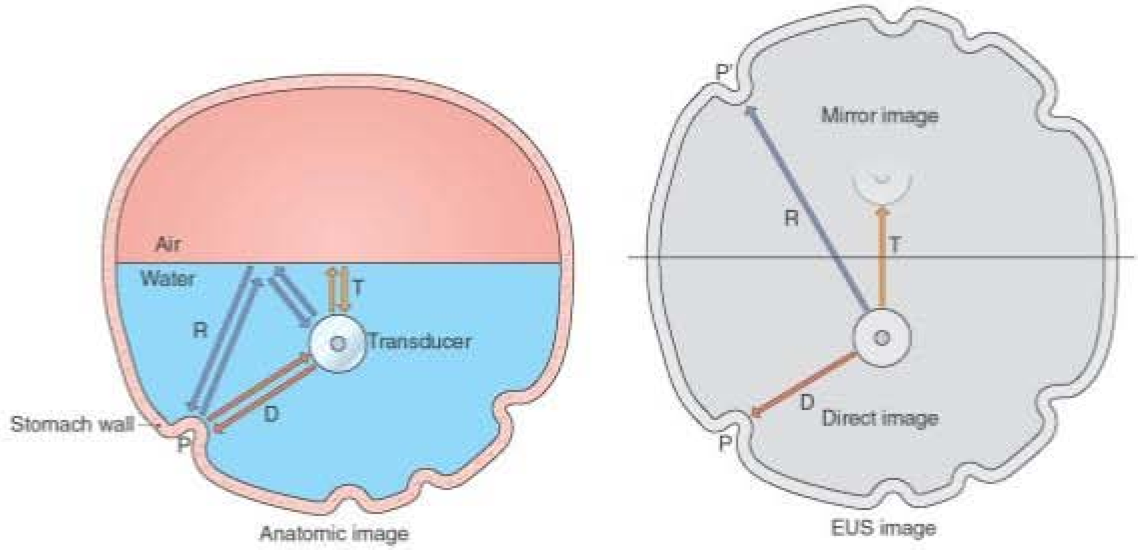

FIGURE 1-19 Reflection from an air-water interface produces a mirror image artifact. Because of the large impedance mismatch between water and air, an ultrasound signal that interacts with an air-water interface is reflected almost completely. The figure on the *left* is an illustration of an ultrasound probe imaging the gastric wall with an air-water interface. The path denoted by D directly images location P along the gastric wall. The path denoted by R images location P because of a reflection from the air-water interface. The path T images the transducer because of a reflection from the air-water interface. The figure on the *right* is an illustration of the resulting ultrasound image. The ultrasound processor registers the location of the image by the direction of the transmitted pulse and the time it receives the reflected signal. The processor accurately registers point P, resulting from the reflected signal from path D; however, the signal from path R is incorrectly registered as point P', with a resulting mirror image appearance. In addition, the reflected signal from path T results in shadowing artifact in the mirror image.

FIGURE 1-20 Shadowing artifact *(arrows)* resulting from calcifications in the pancreas.

FIGURE 1-21 Shadowing artifact *(arrow)* resulting from gallstones *(arrowhead)*.

FIGURE 1-22 Acoustic shadowing *(arrowheads)* resulting from refraction from an interface between normal tissue and tumor.

FIGURE 1-23 Anechoic cystic lesion *(arrowhead)* demonstrating enhancement beyond the cyst relative to other structures *(white arrow)* that are of similar distance from the transducer. This artifact is also called [illegible]sion.

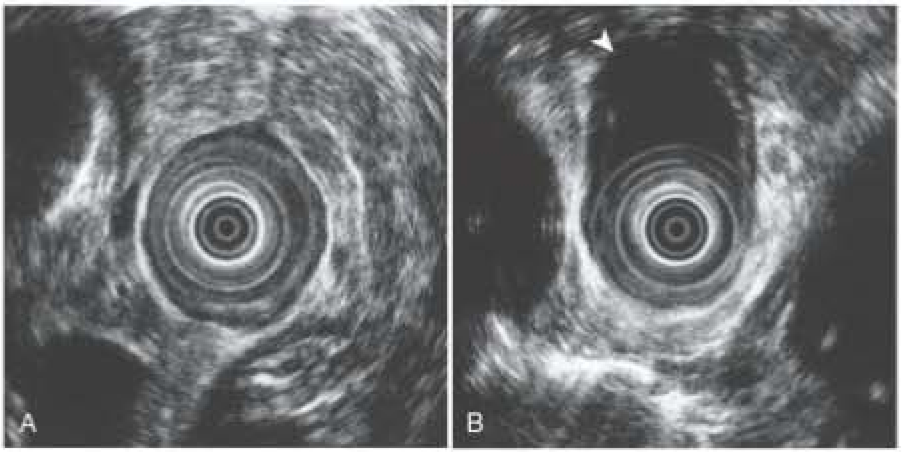

FIGURE 1-24 Tangential imaging artifact. **A**, Normal imaging of a hypertrophic lower esophageal sphincter in a patient with achalasia. **B**, Tangential imaging of the same lower esophageal sphincter (note that the balloon was not inflated during acquisition of this image). The gastrointestinal (GI) tract wall layers are distorted and are not uniformly thick circumferentially, a finding suggesting that the transducer is not imaging a normal GI tract wall. As a result, areas of abnormal thickening are noted on imaging and can give the incorrect appearance of a tumor in the GI tract wall *(arrowhead)*.

obscure any side lobe reflections. However, during imaging of an anechoic structure, the reflected ultrasound energy from a side lobe can be of sufficient intensity to yield a detected signal that is then interpreted by the processor as an on-axis reflection.[10] A side lobe artifact is recognized when the hyperechoic signal does not maintain its position within an anechoic structure such as a cyst or the gallbladder. It may be misinterpreted as sludge in the gallbladder or a mass within a cyst.[9] Figure 1-26 is an image of a side lobe artifact within the gallbladder. Repositioning of the transducer causes the artifact to disappear.

Endoscopic Ultrasound Elastography

Elastography is an ultrasound-based method for evaluation of tissue "hardness," that is, the change in tissue dimensions (strain) arising from an applied force. This concept is closely related to palpation, which physicians have used for centuries to detect pathology associated with higher tissue "stiffness." There are multiple parameters describing tissue elastic properties, including bulk modulus (Equation 1.3, Table 1-1), which describes the change in *volume* of the material in response to external stress. As seen from Table 1-1, bulk modulus only varies by no more than 15% among different tissue types. However, palpation elicits different elasticity parameters—Young's modulus and/or shear modulus, which represent the ratio of tissue *displacement* (or strain) in a certain direction (longitudinal or transverse) to the applied stress. The elastic moduli of normal soft tissues are known to vary as much as four orders of magnitude and are elevated by the pathologic changes, such as fibrosis, by up to two orders of magnitude, with benign tumors being generally softer than malignant tumors.[11]

In elastography, stress is applied to tissue either externally (e.g., vibration, manual pressure, or balloon inflation in the case of transluminal examination) or internally (e.g., by vascular pulsations and respiratory motion). The resulting strain is measured using ultrasound as illustrated in Figure 1-27, showing B-mode images are recorded before and after the application of stress. Each of the B-mode line scans recorded before and after compression is then analyzed using cross-correlation techniques to extract the in-depth strain distribution. These strain line scans are then combined into a

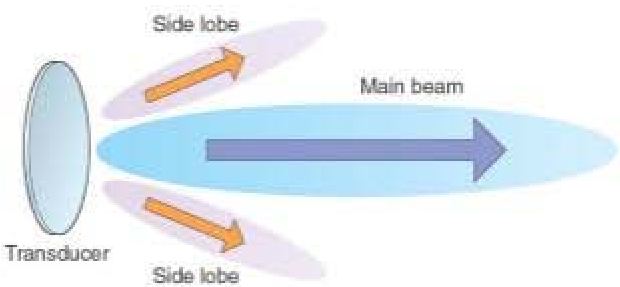

FIGURE 1-25 Side lobes represent secondary projections off-axis from the main beam. Side lobes have lower intensities than the main beam, but they can still produce back-reflected signals from the tissue of sufficient intensity to be detected by the transducer. However, the transducer assumes that all back-reflections originate from the main lobe. Therefore image artifacts can result from side lobe projections.

FIGURE 1-26 Side lobe artifact identified in the gallbladder *(arrow)*. Repositioning of the transducer results in disappearance of this signal.