CHAPTER 1

Introduction to Diagnostic Ultrasound

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A Brief History on the Invention of Clinical Ultrasound

In the year 1790, an Italian physiologist named Lazarro Spallanzani was fascinated with bats and their ability to navigate effortlessly through complete darkness. He marveled at the accuracy with which they could catch their prey without light. Among his numerous experiments with blindfolded bats, he had discovered that they could still maneuver their surroundings without difficulty. He then plugged their ears. Upon observing that their flight became severely erratic, he hypothesized that it was the bat's sense of sound, and not their sight, which provided their primary mode of navigation. In his own words, he stated, "The ear of the bat serves more efficiently, or at least for measuring distances" [1]. He determined from his experiments that bats must emit ultrasonic waves that are inaudible to human ears, and that it was the echoes from these waves bouncing off objects that allowed bats to sense distance, direction, and ultimately their own location in space. Today, we credit Spallanzani as the first to study anything related to ultrasound physics [1]. Echolocation at the time was considered an anomaly and was met with great skepticism by his fellow scientists. Two Harvard students, Robert Galambos and Donald Griffin, would prove his theory over a century later [1].

In 1877, 86 years after Spallanzani's experiments with bats, Jacque and Pierre Curie observed that putting pressure on various crystals created electricity and thus are credited with the discovery of piezoelectricity [2]. It was this discovery that allowed ultrasound transducers to become a reality. In 1915, a physicist by the name of Paul Langevin would use this technology to create the "first transducer," which could detect objects, including submarines, at the bottom of the ocean that was used during World War I, and further developed for use during World War II [3]. These developments laid significant groundwork for the use of ultrasound in medicine. In 1942, Karl Dussik transmitted an ultrasound beam through the human skull with the hope that he could detect brain tumors [4]. He was credited with being the first to use sonography to make a medical diagnosis. After 1951, developments in breast ultrasound imaging and broad field view B-mode sonography demonstrated the feasibility of the application of handheld device and compounding technique in real-time B-mode sonography [5, 6].

In 1958, Ian Donald and colleagues were the first to incorporate this technology into the medical field of obstetrics and gynecology [7]. From there, ultrasound technology made great strides in the field of science, especially in screening for fetal anomalies.

Ultrasound technology has permanently taken a sophisticated and invaluable role in medical diagnostics and treatment. It has utility in clinical and diagnostic settings, such as physical examinations, procedural guidance, and the monitoring of treatment efficacy [8]. The clinical application of ultrasound provides imagery and mechanisms for detecting movement and flow in vascular, organ, and musculoskeletal systems in the human body. Because ultrasound does not emit harmful ionizing radiation, it is ideal for frequent follow-up examinations [8].

Ultrasound is operator dependent. Knowledge of anatomy, pathology, ultrasound principles, ultrasound artifacts and bioeffects, and patient clinical information are all important for operators to perform and interpret ultrasound examinations adequately in clinical patient care [8, 9].

The Basic Physics and Instrumentations of Ultrasound

Sound, Waves, and the Range of Human Hearing

Sound is a form of mechanical energy that travels through matter and is created by a vibrating source. This mechanical energy produces a phasic pattern of oscillations, which is characteristic of waveforms. Unlike light waves, sound waves require a medium to propagate [8]. Individual particles of the medium exhibit particle motion parallel to the direction of traveling longitudinal sound waves. This allows for acoustic propagation of mechanical energy in a domino-like fashion [10].

Acoustic Parameters

Sound waves, like all waves, are defined by the following parameters: frequency (MHz), wavelength, amplitude, power, and intensity [8]. While frequency and amplitude are determined by the source, the propagation speed of a sound wave is determined by the medium, and frequency and wavelength are related through the propagation speed [8, 10].

The spectrum of sound may be described in relation to the frequency range of human hearing. In most humans, the range of hearing lies between 20 and 20,000 Hz [10]. Sound frequencies below 20 Hz are referred to as infrasound, while sound frequencies above 20,000 Hz are referred to as ultrasound. Medical ultrasound typically ranges from 1 to 50 MHz (much higher than that of human hearing). Frequency (f) is the number of oscillations of a wave or particle per second and is independent of the medium. It is expressed in Hertz (Hz). These oscillations may be pictured as the cyclic compressions and rarefactions on a spring. As you push one end of the spring, the energy is propagated longitudinally along the coils, similar to that of molecules of a medium propagating a sound wave. The amplitude of the wave at a given moment of time is the distance from the mean position of the particle to the actual position of the particle. The peak amplitude is the distance between the mean and the maximal displacement. The period (T) of the wave is the time it takes for each oscillation to occur, expressed in seconds (s), and is the inverse of the frequency [8, 9].

Wavelength (λ) is the distance traveled in one oscillation and may be represented as one complete cycle or the distance from compression to the next compression or rarefaction to the next rarefaction on a longitudinal wave. For a sound speed of 1600 m/sec, the wavelengths in ultrasound application vary from 0.8 mm at 2 MHz to 0.08 mm at 20 MHz. Frequency and wavelength are inversely related; as frequency increases, wavelength decreases [8–10].

The Interactions Between Ultrasound and Tissues

When ultrasound waves are sent into the body, they interact with various media (e.g. air, bone, and tissue). Some of these waves are scattered, and some are absorbed and dissipated mostly as heat [10]. The echoes that are reflected or scattered back to the transducer are translated from electrical information to a sonographic image by the ultrasound machine. The amplitude of echoes returned (backscatter) is determined by a property of tissue called the acoustic impedance along with any effects associated with directional scattering and losses along the returning path [8]. This intrinsic property of a medium is determined by the density of the medium multiplied by the propagation speed of the sound wave [11].

Acoustic impedance is a measure of an individual particle's resistance to the mechanical energy contained in the acoustic field (Table 1.1) [11]. This resistance is proportional to both the density and propagation speed of the wave. The importance of this concept comes into play when determining the amplitude of the echoes returned to the transducer by two media with different acoustic impedances.

TABLE 1.1 **Acoustic impedance values of various media.**

ACOUSTIC IMPEDANCE OF VARIOUS MEDIA	
MEDIUM	**ACOUSTIC IMPEDANCE (MRayl)**
Air	0.0004
Fat	1.38
Water	1.48
Kidney	1.62
Liver	1.65
Bone	7.8

Source: Data from [11].

tWhen ultrasound pulses hit a large smooth surface of two media of different acoustic impedances, mechanical energy is reflected back to the transducer. This is called specular reflection, and the intensity of the echoes reflected is related to the difference in the acoustic impedances of the media [8, 11].

Reflection specifically occurs at tissue interfaces, which are junctions of tissues with different acoustic impedances [8, 9]. Therefore, if the sound waves pass through tissues of similar impedances (homogeneous tissues or fluid), few or no echoes will be produced, and the area being imaged will look dark or anechoic. Sound waves passing through tissues with large differences in acoustic impedances produce larger amplitude echoes and, therefore, produce bright or hyperechoic appearance in the image. It is the amplitude of the reflected sound that determines the relative brightness of the image. Air has a much lower acoustic impedance compared to soft tissues or bone. Therefore, when an ultrasound beam hits an interface between air and tissue, there is a large reflection of the ultrasound, and very little ultrasound is transmitted to the tissue distal to the air [8, 9].

Snell's law, which is often used to describe light waves, can also be used to describe the reflection and refraction interactions of ultrasound waves at an interface (Fig. 1.1). As ultrasound hits a linear interface at 90°, the echo from that interface will travel back to the transducer. In contrast, if it hits a linear interface at any angle less than 90°, the echo may miss the transducer and escape detection [10]. It will instead reflect at an angle equal to the angle of incidence, similar to light reflecting off a mirror. Refraction refers to a change in the direction of sound transmission after hitting an interface of two tissues with different speeds of sound transmission [11]. For example, when a beam hits an interface at an angle other than 90°, the portion that continues forward changes direction (bends). Refraction is an important cause of incorrect localization of a structure on an ultrasound image, which will be discussed in greater detail under artifacts.

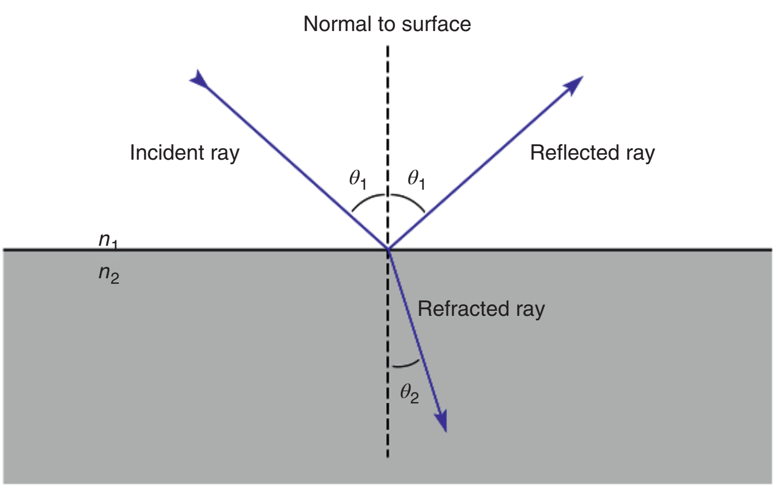

FIGURE 1.1 This illustration explains ultrasound wave propagation at an interface (Snell's law). Waves traveling in the medium n_1 and hitting the interface of n_1 and n_2 at 90° produce echoes that travel back toward the transducer. Waves hitting at <90° (θ_1) produce waves that are reflected at the same angle and may escape detection. The portion of these waves that then travel into n_2 will be refracted to travel along a new direction (θ_2) that depends on the relative values of n_1 and n_2.

Scattering occurs when ultrasound pulses encounter rough and irregular tissue structures [11]. The echoes are reflected in a wide range of angles, which reduces their intensity. When this scattering is caused by structures in the tissue that are smaller than the wavelength of sound, we get a pattern in the ultrasound image termed speckle, resulting from interference among these scattered signals. Speckle gives a textured or grainy appearance to most biological tissues and organs on imaging. The use of multiple beams coming from different directions helps to address this issue by averaging out speckle in the image, which in turn yields reduced noise and improved image contrast and subsequently better delineation of anatomical features, such as a focal lesion in the liver.

Any time an ultrasound wave is sent through a tissue, there is some reduction of intensity due to attenuation. There are two significant factors that contribute to attenuation. One is the scattering by the mechanisms described above. The other is absorption, which is mechanical energy dissipated typically as heat. This conversion of energy into heat is the most important contributor to attenuation in ultrasound [8, 10].

Higher frequencies have shorter wavelengths and thus give the operator higher axial (and lateral) resolution images, allowing for better discrimination between tissues. This is why higher-frequency waves are more helpful for imaging superficial structures. However, one must consider depth when selecting a given frequency for ultrasound, as higher-frequency waves are also attenuated more for a given distance compared to lower-frequency waves [11, 12]. Lower-frequency ultrasound waves have longer wavelengths that penetrate deeper into tissue due to a decreased cumulative attenuation but provide the operator poorer discrimination between tissues. Optimizing the choice of transmitted frequency for a variety of clinical scenarios is one of the fundamental challenges faced in ultrasound. Adjusting the amplitude of the transmission could obtain a deeper penetration of tissues, but regulations and safety issues limit the maximum pressure amplitudes that can be transmitted into tissue. This is where coded excitation can come into play. Coded excitation is a technology in ultrasound that allows the user to increase the signal-to-noise ratio for a given frequency and thus increase imaging depth without increasing the amplitude. Coded excitation methods use transmitted pulses that are much longer than typical imaging transmission pulses. The increased temporal energy in the pulse improves signal-to-noise ratios, but additional signal processing is needed to prevent loss of axial resolution. Axial resolution can be recovered by a matched filter process (circuit or algorithm), specific to the code embedded into the transmitted pulse [13].

The relationship between frequency and wavelength is connected through the sound speed of the medium, that is, the product of frequency and wavelength is equal to the propagation speed of the wave (m/s). As stated previously, the frequency of a wave is determined by the transmitting source and therefore must remain constant in the equation. While frequency is constant, the propagation speed will change depending on the density and stiffness of the tissue through which the ultrasound waves travel. As a property of the medium, the sound speed then sets the wavelength for the given frequency. Based on the relative compressibility of the medium, the speed of ultrasound through gases is low, liquids high, and solids the highest [8, 10, 11]. Similarly, the propagation speed is much faster in bone and fluid-filled organs than in air (Table 1.2) [8]. A balance of all these factors (propagation velocity and associated wavelength, imaging depth, etc.) plays a role in obtaining a high-quality image.

$$v = f\,\lambda$$

Velocity (v) is equal to the frequency (f) multiplied by the wavelength (λ).

TABLE 1.2 Propagation speeds of ultrasound through various media.

DENSITY AND PROPAGATION SPEED OF VARIOUS MATERIALS		
MEDIUM	**DENSITY (kg/m^3)**	**SPEED (m/s)**
Air	1.2	331
Fat	925	1450
Water	1000	1540
Muscle	1050	1547
Bone	1400–1900	4080

Source: Data from [8].

Another factor that affects the amount of attenuation is the distance traveled by the wave. The longer the ultrasound waves must travel and the higher their frequency, the greater the cumulative attenuation. The type of medium (e.g. tissue) also affects the amount of attenuation. Water and blood attenuate ultrasound by a barely appreciable degree, while cortical bone presents a virtually impermeable barrier due to its impedance mismatch to tissue and its attenuation (Table 1.3). Ultrasound equipment typically compensates for nominal tissue attenuation and automatically increases the amplitude (and thus overall brightness) of the signal in deeper areas of the image, commonly referred to as time-gain compensation (TGC). The attenuation experienced by waves in various media is quantified using the attenuation coefficient (α), having typical units of dB/cm-MHz, showing the increasing amount of attenuation with distance traveled and with increasing ultrasound frequency. This coefficient helps in assessing the amount of attenuation that ultrasound waves experience in certain media. The ratio of the two echo

TABLE 1.3 Attenuation coefficients of various media at a frequency of 1 MHz.

ATTENUATION COEFFICIENTS OF VARIOUS MEDIA ($f = 1$ MHz)	
MEDIUM	**(α = dB/cm)**
Blood	0.2
Fat	0.48
Liver	0.5
Muscle	1.09
Air	1.64 (20°C)
Bone (cortical-trabecular)	6.9–9.94

Source: Culjat et al. [14] and Christensen et al. [15].

amplitudes (intensities) is often expressed on a logarithmic scale, decibels (dB) [14, 15]. For example, if the ratio of two echo amplitudes is two, then expressed in dB, this would be $20 \log(2) = 6\,\mathrm{dB}$. In terms of intensities, a ratio of 2 would be $10 \log(2) = 3\,\mathrm{dB}$.

Transducers and Sound Beams

Ultrasound (US) technology generates imaging through the conversion of energy between its different forms. The change of mechanical to electrical energy is referred to as the piezoelectric phenomenon, while the inverse conversion of electrical to mechanical energy is known as the reverse piezoelectric phenomenon [10].

To produce waves, an external electric voltage stimulates crystalline material within transducers to undergo vibrational and compressive kinetic movements capable of emitting US waves [16]. Therefore, the production of a US beam is accomplished using the reverse piezoelectric phenomenon. Contrarily, incoming beams that are reflected from tissues hit the crystalline material, causing a kinetic change that generates an electric signal, demonstrating the piezoelectric phenomenon in the detection of incoming acoustic signals.

Structural components of transducers optimize the piezoelectric effect to produce clear ultrasonographic images. Transducers typically consist of the piezoelectric plate, a matching layer, and a backing layer. A lens may also be present in transducers. Each transducer may have a varying number, size, shape, and arrangement of transducer elements used for different applications [10].

The piezoelectric element of US transducers may consist of a synthetic ceramic material, such as lead zirconate titanate (PZT). The thickness of the plate determines the frequency of vibration. Appropriate thickness allows waves to resonate under the constructive interference wave theory, thus reinforcing each other. Resonance is seen when the plate thickness is one-half the wavelength of the desired frequency [13]. Electrical activation of thinner plates produces higher frequencies [16].

The backing layer dampens the oscillation of the piezoelectric plate to reduce the duration of the ultrasound pulse. The acoustic insulator of transducers localizes plate vibrations and prevents energy from flowing into the transducer housing. An acoustic matching layer is seen anterior to the piezoelectric plate and functions to match the high impedance of the plate to the relatively lower impedance of the tissue and to place outgoing waves in phase so that resonance occurs [10]. This improves transmission into soft tissues as well as reception from tissues. Generally, improved reception is more crucial since there is an upper limit of power transmitted into soft tissue. The more acoustic signal the matching layer lets through to the piezoelectric plate, the better the resulting image [16].

The transducer description given here is a classic construction scenario. There are many new technologies that have been applied in transducer construction such as composite materials for the piezoelectric layer and even replacing the piezoelectric with alternative ways of producing and receiving ultrasound such as the capacitive micromachined ultrasound transducer (CMUT). These are beyond the level of discussion presented here, and readers interested in further details are referred elsewhere [17, 18].

Ultrasound Beam Shape

Once the ultrasound leaves the transducer, the beam that is produced has multiple characteristics that can be viewed and studied. These are discussed below.

Non-focused Source

A planar disc transducer, sometimes referred to as a single-element (planar) transducer, is a non-focusing source from which an ultrasound beam is emitted. When emitting a (near) continuous wave (CW) at a given frequency, the pathway of the beam can be divided into the near field (Fresnel zone) and the far field (Fraunhofer zone) (Fig. 1.2). The beam width in the near field is approximately the transducer width, or aperture [13]. At the end of the near field comes a point where the beam reaches its last maximum intensity of constructive interference along its axial route (Fig. 1.2). This marks the position at which the beam begins to spread laterally at an angle known as the angle of divergence [13]. Larger transducers and lower wavelengths minimize the angle of divergence, while smaller transducers and higher wavelengths increase the angle at which the beam spreads. The near field is influenced by the same properties, such that a longer near field will result in a smaller angle.

Focused Source

Transducers with focusing capability are commonly used in medical imaging and produce beams that exhibit different wave properties than the unfocused transducers discussed above. As the beam leaves the transducer, it gradually decreases in width until it

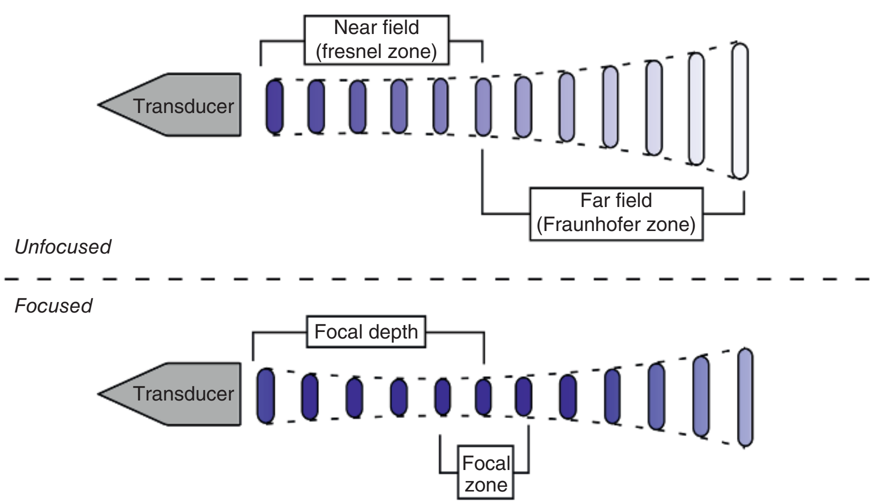

FIGURE 1.2 General shape of US beam (top) unfocused transducer: the near field is of equal width as the transducer and begins diverging after the natural focal point, becoming the far field. The angle of divergence represents the lateral spread of the beam once it passes the focal point. (Bottom) Focused transducer: beam width gradually decreases as the beam leaves the transducer, reaching its minimal width at the focal point. After the focal point, the beam diverges similarly to the unfocused US beam.

reaches the focus, or its narrowest point. The focus also has the highest intensity of the entire beam and produces the highest spatial resolution. The focal length is the distance from the transducer to the focal point. Once the ultrasound beam reaches its focal point, it begins to diverge in a similar pattern seen in unfocused beams (Fig. 1.2).

Alternative Beamforming

It must be noted that the descriptions above are for the traditional approaches to transmitting ultrasound. There have been several advances made that include the use of planar transmitters, similar to non-focused source described above, that are combined with beam steering from multiple angles. Receiving echoes from these different transmits can be combined with receive beamforming for high frame rate ultrasound imaging. For further information, the reader is referred elsewhere [19].

Image Formatting

A variety of transducer types have been developed to optimize imaging requirements based on the examination performed. The clinical application of transducer formats will be discussed thoroughly in the following chapters.

As the application for which the ultrasound imaging varies, so does the format (Fig. 1.3). Linear array transducers (Fig. 1.3a) are composed of an array of elements oriented in a straight line, typically producing a rectangular image as the beam is transmitted into tissues perpendicular to the array. Linear array transducers are commonly used in vascular, musculoskeletal, and superficial structure ultrasound examinations. Curvilinear transducers (Fig. 1.3b) are similar to linear designs, but the elements are on a slightly curved surface. While both transducers offer a wide field of view, the curvilinear transducer's field of view expands with increasing tissue depth [13]. These transducers are useful when a broad image is required that is not obstructed by bones or gas. This may include evaluations of the abdomen, pelvis,

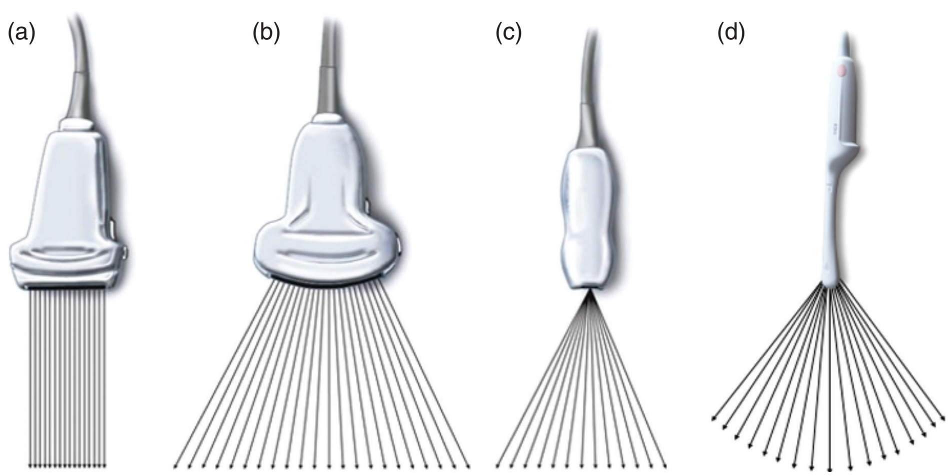

FIGURE 1.3 Transducers. Transducer formats for linear array (a), curvilinear array (b), phased array (c), and endocavity array (d).

obstetrics, and gynecological organs [16]. Phased array transducers (Fig. 1.3c) with smaller transducer footprints are useful in performing transthoracic imaging of the heart. Endocavitary transducers (Fig. 1.3d) are inserted into body cavities or surgical openings to allow transducers to be placed closer to target tissues. This allows operators to scan these structures with higher frequencies, improving spatial resolution and diminishing image distortions and artifacts [13]. Note that combinations of these imaging approaches can be made, such as using some beam steering with linear arrays to create trapezoidal image formats.

Use of 3D/4D imaging has expanded the use of ultrasound imaging by allowing manipulation of images into an alternate imaging plane, the coronal plane, and for displaying a three-dimensional image. Both of these imaging methods can provide additional diagnostic information for clinical use. 3D imaging can be conducted with a curvilinear array and endocavity transducers that are rotated slowly along multiple 2D image planes to compose a 3D image [16]. An alternative method of providing a 3D image includes the use of a 2D transducer that is steered along the traditional lateral and elevational directions [16].

Resolution and Image Characteristics

Spatial resolution allows operators to distinguish between two objects located at two different positions in space. Axial resolution is the ability to identify such objects at different positions along the axis of the ultrasound beam, one behind the other [20]. Axial resolution is improved with higher frequencies due to shortened wavelengths and when using shorter pulse lengths [10]. It does vary to some degree with tissue depth as the bandwidth of the pulse changes with depth due to frequency-dependent attenuation.

Lateral resolution differentiates reflectors that are located side-by-side, perpendicular to the US beam [20]. Lateral resolution is improved with narrower beam width, proximity to focal plane, higher frequency, and higher scan line density [10].

As spatial resolution improves at higher frequencies, it is subject to greater attenuation [16]. Lower frequencies must, therefore, be used to reach deeper tissues, with the associated reduction in spatial resolution. Ultrasound technology incorporates broad-bandwidth technology to optimize image quality and balance the benefits seen in high and low frequency beams.

Bandwidth is defined as the range of frequencies that a transducer can produce and detect. Pulsed wave ultrasound can produce a wider range of frequencies, and short pulses have larger bandwidths. By producing beams with a larger bandwidth, a greater spectrum of reflected echoes can be detected from tissues insonated, ultimately improving spatial resolution [21]. In simple terms, larger bandwidth allows the production of shorter pulses, which allows for better separation of echoes from closely spaced targets.

The capability of differentiating the time over which events occur is known as temporal resolution [10]. The rate at which images are produced influences temporal resolution, with higher frame rates being more capable of distinguishing moving objects in time. Increasing the number of scan lines in an image can improve lateral resolution but will decrease frame rates. Decreasing frame rates will worsen temporal resolution [16]. This is also true in deeper structures, where increased depth requires lower frame rates due to longer beam travel time and worsening temporal resolution.

The distinction of echo amplitudes, seen by grayscale differences in displayed pixels, is referred to as contrast resolution [16]. These differences in contrast are used to identify objects in an image, and the combination of spatial and contrast resolutions limits the size of objects that can be seen [13]. Background noise may sometimes be seen from electrical sources and/or speckle, sometimes resulting in difficulty distinguishing true echoes and diminishing contrast resolution. Noise can be reduced by averaging several images over time or with spatial compounding [16].

Display Modes

Amplitude Mode (A-Mode)

One of the earliest forms of medical ultrasound was the amplitude mode, or A-mode. As echoes return from tissues, the amplitude of the received signal on the transducer was portrayed on a cathode ray oscilloscope (CRO). On a CRO, the vertical axis corresponds to the amplitude of the returning echoes while the horizontal axis corresponds to the time scale for the echo return (Fig. 1.4a). As the strength of echoes increases, amplitude rises as well. Increased time along the horizontal axis implies

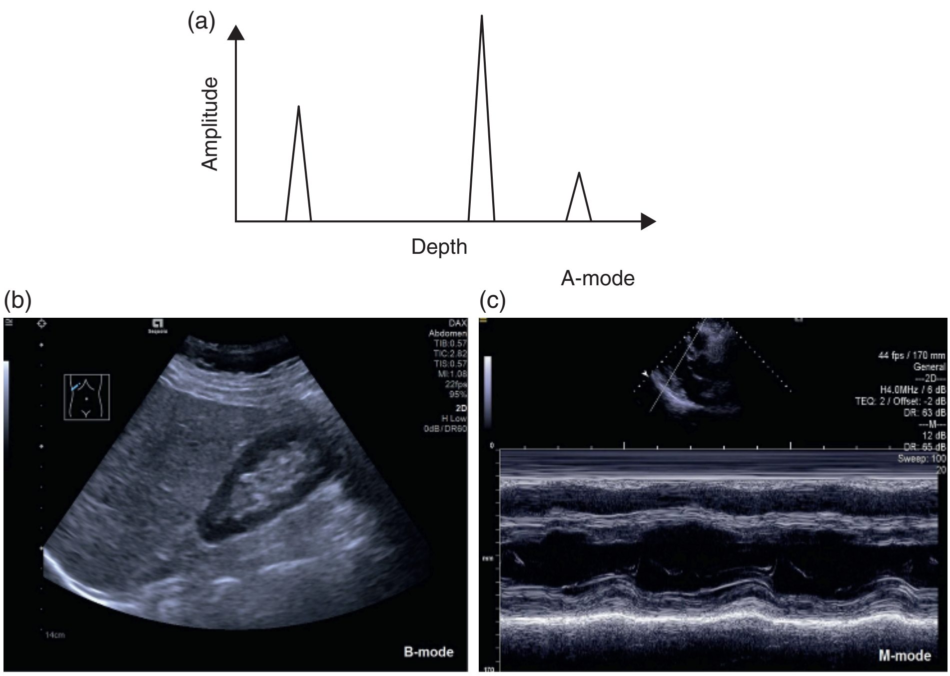

FIGURE 1.4 Ultrasound display modes: (a) Amplitude mode (A-mode) represents the amplitude of returning echoes projected along the x-axis at varying times along the y-axis corresponding to tissue depths. (b) Brightness mode (B-mode) is a common 2D image display for anatomic structures based on the intensity of gray pixels (related to echo strength) and depth of the tissue imaged. (c) Motion mode (M-mode) uses a user-selected line through the B-mode image (top half) and portrays the movement of structures along that line over time (bottom half).

that the echoes travel further distances and are located deeper within the body and further from the transducer. A-mode produces a one-dimensional image of returning echoes. The image produced when using A-mode ultrasonography is a simple time trace of this signal [10, 20].

Brightness Mode (B-Mode)

Returning echoes are portrayed as gray dots of varying brightness corresponding to echo amplitude (Fig. 1.4b). Strong echoes produce increased brightness. Weak echoes produce dark regions in the image as they are returning from nonreflective tissue. Vertical direction within the image corresponds to the distance between the reflector and the transducer. The use of multiple scan lines produced from the transducer's surface is properly positioned horizontally to generate a 2D image of the structures through which the beams pass. Most ultrasound machines display images with up to 256 shades of gray. This is now a common mode used for diagnostic imaging [20].

Motion Mode (M-Mode)

M-mode generates an electronic trace of a moving object lying along the path of a single ultrasound beam. The transducer is placed in one fixed position in relation to the moving structure. If the distance between the transducer and the reflector changes due to movement, the corresponding image will display that change as a function of time (Fig. 1.4c). Stationary structures will remain in place, while moving structures will move along the vertical axis of the screen as a function of time (horizontal axis). Among many of its uses, M-mode is often displayed simultaneously with B-mode to study cardiac valves, calculate fetal heart rate, and evaluate lung motion [20].

Other Imaging Modes

There are other ultrasound imaging modes that have been developed over time such as elastography that have not been covered here. The reader is referred elsewhere for additional information [22].

Real Time and Freeze

US machines can image dynamic movement of tissues with their capacity to produce high frame rates for real-time imaging. This is useful in observing moving organs and serves as a major advantage of US imaging when compared to diagnostic tools such as CT or MRI.

The freeze function allows the operator to interrupt the real-time acquisition at a given moment and analyze the features of the image and store it for further investigation. When frozen, the transducer usually stops transmitting US pulses [10]. When the image is frozen, the most recently collected images are temporarily stored in a series known as a cine loop. The entire cine loop or portions of it can then be archived.

Doppler Ultrasound

Doppler ultrasound can be used to detect motion, such as blood flow [13]. Moving components within the body function serve as reflectors for sound emitted by the transducer. Their change in velocity alters the frequency that returns to the transducer. This change in frequency is used in continuous wave (CW) ultrasound systems to determine the Doppler shift [16].

Wave fronts from targets moving toward the transducer are more tightly packed, resulting in a higher-frequency wave than the one emitted. In contrast, lower frequencies are seen when targets are traveling away from the transducer [16]. The magnitude of the Doppler shift frequency is proportional to the relative velocity between the source and the observer [13]. Therefore, as the velocity of moving structures such as red blood cells increases, the Doppler shift also increases.

The angle of approach by the transducer impacts the Doppler shift [16]. The angle between the direction of sound emission and the direction of blood flow should be less than 60° (Fig. 1.5) to reliably measure Doppler shift. Flow velocity (Doppler frequency) is reduced in relation to the increase in cosine of the Doppler angle, and the cosine of angles greater than 60° varies rapidly [13].

$$\text{Doppler shift} = \frac{2 \times \text{reflector speed} \times \text{incident frequency} \times \cos(\text{angle})}{\text{Propagation speed}}$$

According to the above equation, Doppler shift is directly related to the reflector speed (tissue velocity), the incident frequency (frequency of the transmitted ultrasound), and the cosine of the angle of incidence. Propagation speed is the speed of sound in the medium and is inversely related to the Doppler shift [23].

Doppler signals are mostly applied when studying blood flow but can also be used when analyzing tissue motion [16]. Blood signals are most commonly of low amplitude and high frequency shift. Tissue signals appear as high amplitude and low-frequency shifts due to the relatively slower velocities when compared to blood flow in vessels of appreciable size, for example, capillary velocities are too small for most Doppler systems to detect separately from tissue motion. The use of a wall filter

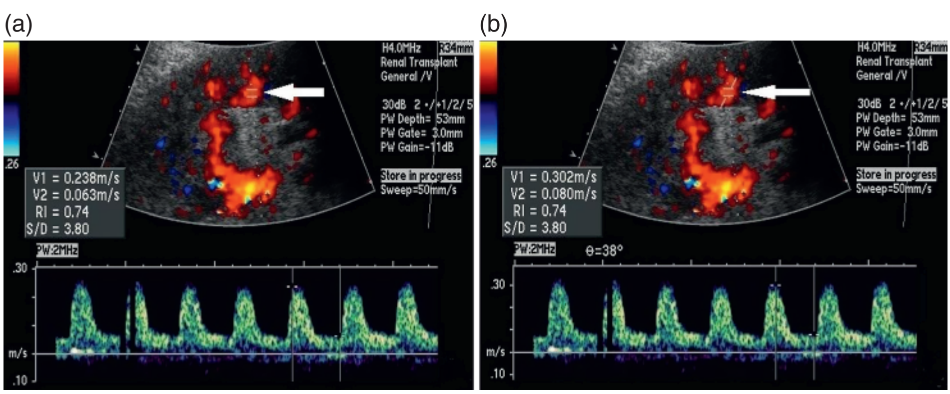

FIGURE 1.5 The operator should use Doppler angle correction when measuring flow velocity using spectral Doppler. There is a difference in flow velocity measured without (a, white arrow) and with (b, white arrow) using Doppler angle correction (0.238 versus 0.302 m/s).

allows the ultrasound machine to distinguish Doppler produced by blood flow compared to the surrounding tissues [10].

Important Note: While the Doppler effect described above is the origin of the Doppler modes seen on ultrasound machines, the actual measurement methods used to detect motion may not detect the actual frequency shift. In fact, the actual Doppler effect as measured by CW ultrasound is rarely used. Other "Doppler" modes use phase shift. This can be easily seen by the fact that there can be aliasing as will be described below. The Doppler effect itself does not alias. However, we commonly retain the terms color Doppler and pulsed Doppler even though the methods rely on phase shift methods. Perhaps more properly, these should be, and sometimes now are, termed color flow and pulsed wave [24].

Spectral Doppler (Pulsed Wave) and 2D Color Flow Images

Velocity changes within a specific area can be analyzed using spectral Doppler, also called pulsed wave [23]. Velocity information that is read from a single location is represented as a frequency shift (vertical axis) as a function of time (horizontal axis) [13]. Positive or negative on the vertical axis corresponds to the direction of the Doppler shift, while the grayscale (sometimes colorized) shows the strength of the Doppler signal (Fig. 1.6).

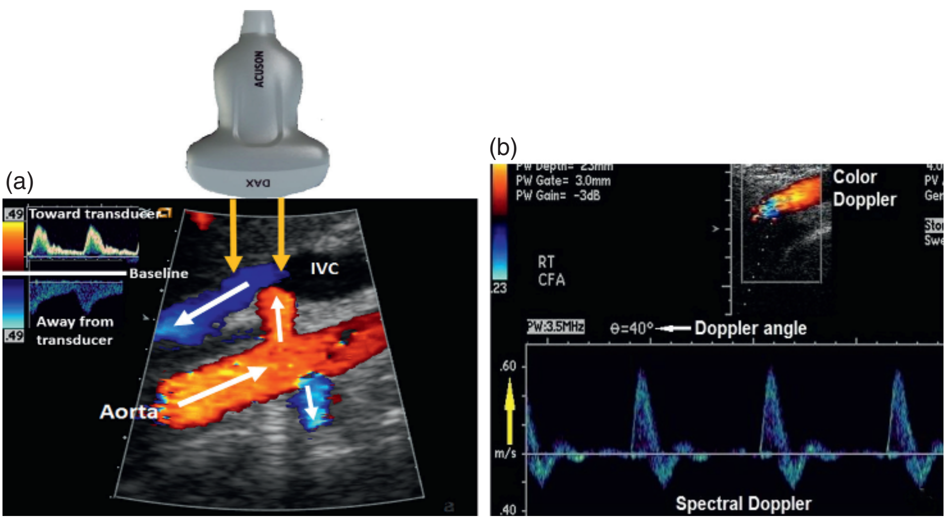

FIGURE 1.6 (a) Color flow imaging representing the mean Doppler frequency and flow direction is superimposed on the B-mode image. Doppler displays positive (red color and spectrum above the baseline) representing flow toward the transducer and negative Doppler frequency shifts (blue and spectrum below the baseline) representing flow away from the transducer at a specific location over time. White arrows show the direction of blood flow, and yellow arrows indicate the nominal direction of transmitted sound beams. IVC, inferior vena cava. (b) Vertical distance in the spectral trace corresponds to the Doppler shift (flow velocity) scale (yellow arrow). A Doppler angle (white arrow) less than 60° is recommended to keep the direction of the transmitted sound beam as parallel to the direction of blood flow as possible, which can better measure the flow velocity. RT, right; CFA, common femoral artery.

The Doppler signal can be portrayed as a 2D color image superimposed on a B-mode image (Fig. 1.6). Color flow Doppler corresponds to the Doppler shift (actually phase shift) and depicts direction of flow and mean velocities (not corrected for angle) [13]. By convention, blood flow toward the transducer is seen as red and blood flow away from the transducer is seen in blue (although the user can typically reverse this color scale if desired). A complication in color flow Doppler is the potential for aliasing, in which color flow direction information in vessels may be distorted, making accurate determination of blood flow direction difficult [21, 23]. Spectral Doppler can alias as well if the pulse repetition frequency (PRF), also referred to by other names such as Scale, is not set properly. Power Doppler is similar to color flow Doppler but displays the power of the Doppler signal rather than its velocity [25]. Therefore, it does not have the same aliasing issue while having the advantages of being less dependent on Doppler angle and providing increased sensitivity.

Continuous Versus Pulsed Wave Doppler

A continuous wave (CW) Doppler uses two transducer elements, with one transmitting a CW ultrasound signal and the other receiving the Doppler shifts. The CW ultrasound allows for highly accurate Doppler measurements in high-velocity flow such as aortic valve stenosis. A disadvantage of CW Doppler is the range ambiguity since there is no arrival time information as with pulsed ultrasound [23]. Pulsed wave (PW) Doppler is characterized by a single transducer that serves in both sending and receiving Doppler signals. Compared to CW Doppler, pulsed wave Doppler has improved control over the specific area being interrogated but is limited in Doppler accuracy and in high-velocity measurements due to aliasing [23].

Artifacts

Artifacts can cause visual misrepresentations and may distort the representation of anatomical structures. Operators must identify artifacts to interpret diagnostic ultrasound images correctly.

A major artifact is acoustic shadowing that can also be used for certain diagnoses. Shadows are formed when beams are unable to pass through a strongly attenuating or reflecting structure [13]. Shadows normally project directly posterior to the structure along the beam axis, typically vertically for linear arrays and radially in phased or curvilinear arrays (Fig. 1.7). Shadowing is formed by large stones, calcifications, and bones, which are caused mainly by sound absorption, refraction, and reflection. Acoustic shadowing can also interfere with lung images due to US scattering at the interface between tissue and air, while the presence or absence of shadowing may also be diagnostic for some conditions.

Reverberations are reflections of multiple repeated pulses and echoes formed by strongly reflecting interfaces [16]. Strong reflective tissues that are relatively shallow are the most common sources of reverberations. As a strong echo returns to the transducer, it may be reflected from the transducer toward the tissue. It then returns to the transducer as a second, weaker echo (Fig. 1.7). This subsequent echo is known as the reverberation. As reverberations are produced from the ultrasound wave traveling to the transducer twice or more, the ultrasound machine registers them as traveling

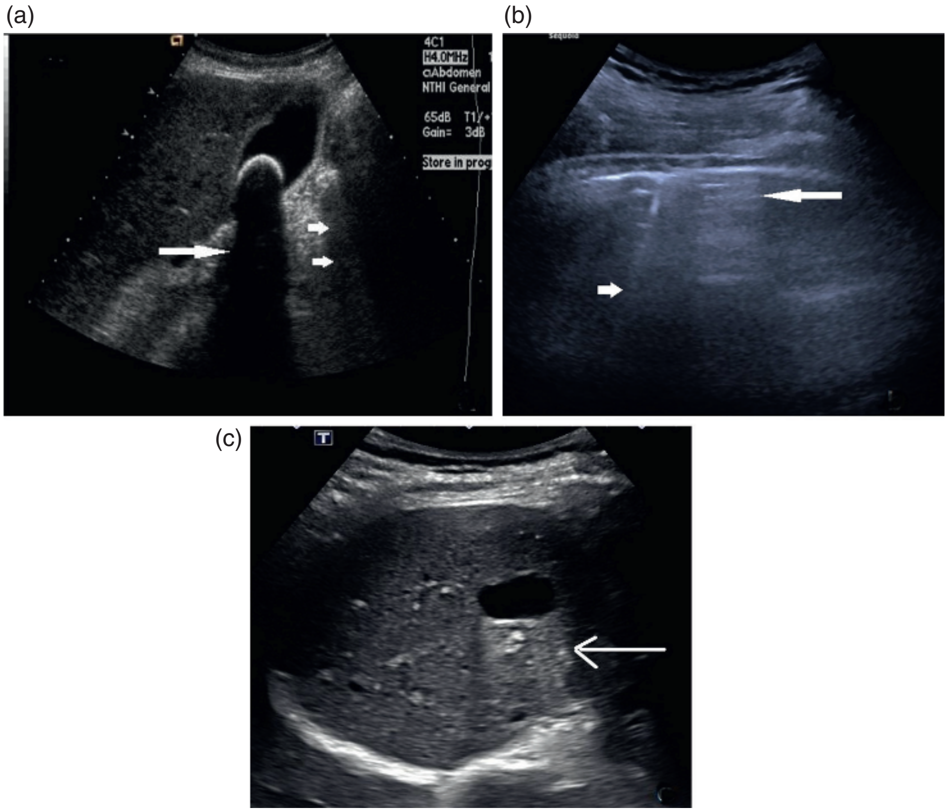

FIGURE 1.7 Common ultrasound artifacts. (a) Shadows posterior to highly reflective gallstones (long white arrow) and bowel gas (short white arrows). (b) Reverberations appear as multiple echoes produced by strong reflecting interfaces (A-line, long white arrow). Comet tails (B-lines, white short arrow) are formed by hyperechoic reflections perpendicular to what is being imaged in the lung ultrasound. (c) Posterior acoustic enhancement (white arrow) is shown as the ultrasound beam travels through an anechoic and low attenuation structure to produce an image artifact observed posterior to a liver cyst.

a further distance and portrays them as objects located deeper than the reflecting surface from which they originate [26].

Reverberations can be identified from authentic echoes by their response to gentle pressure on the skin. A genuine echo will move closer to the transducer at the same rate by which the transducer is pressed deeper against the skin, while the separation of reverberation bands will decrease at twice the rate due to their multiple pathways.

Reverberations are common in images of liquid-filled areas or in dental implants [27]. For example, the anterior bladder wall may demonstrate reverberations due to multiple reflections between it and the transducer, which are readily seen in the anechoic, urine-filled bladder. Two types of reverberations are ring-down and comet-tail artifacts (Fig. 1.7). Ring-down reverberations display a bright vertical line from objects such as an air bubble that travels down the image, while comet tails are hyperechoic trails that are seen in structures such as lung surface irregularities, foreign bodies, and cholesterol deposits in the gallbladder wall. Comet-tail artifacts result from echoes bouncing back and forth within two reflective surfaces of

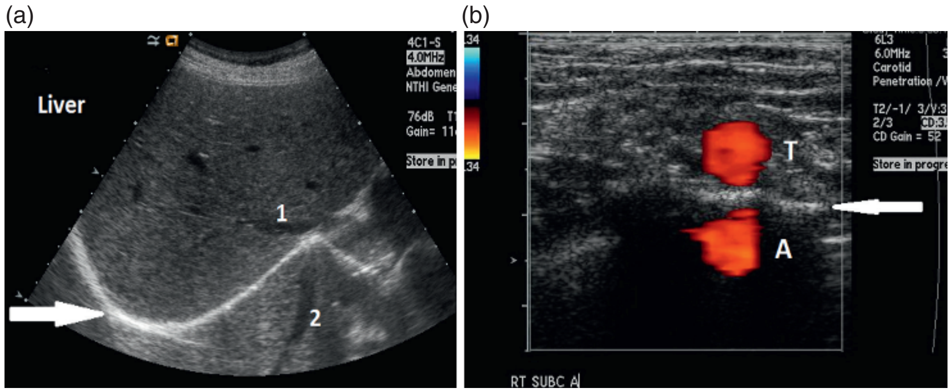

FIGURE 1.8 Images containing true structures and artificial ones appearing as a mirror image adjacent to an echogenic-curved reflector such as the diagram (a, white arrow) and subclavicle bone (b, white arrow). (a) In B-mode ultrasound image, the artifactual liver parenchyma and hepatic vein (2) appear similar to true liver and hepatic vein (1). (b) In color flow imaging, artifactual vessel (A, artifactual vessel) is a mirror to the true subclavian artery (T, true vessel). RT, right; SUBC A, subclavian artery.

an object, but with some energy of each interaction returning to the transducer. As echoes return to the transducer, each new echo produces a weaker signal resembling a comet tail on the US machine when viewed together [26].

Acoustic enhancement (Fig. 1.7c) is also known as posterior acoustic enhancement due to locally increased transmission to the tissue posterior to a structure that contains liquid [26]. As the US beam travels from tissues through weakly attenuating structures such as a cyst or blood vessel, it returns to the transducer with greater intensity than adjacent areas that travel through uniform tissue. This is displayed as structures seen immediately posterior to these structures as having increased echogenicity. The enhancement artifact assists in making the diagnosis of a cystic structure.

Mirror images appear when echoes go through multiple reflections, causing an altered image formation [26]. A beam that hits a bright reflector may hit an object on its return to the transducer. The echo thus takes a longer path to reach the transducer and registers on the machine as a second object that is deeper than the bright reflector. A mirror image is thus created in B-mode (Fig. 1.8a) and color flow imaging (Fig. 1.8b) [28]. In addition, scattering from the object may be sufficiently strong to again reflect from the mirroring surface and return an echo to the transducer along the original beam path. The mirror image can disappear with transducer adjustments, whereas a real object will remain on the screen after transducer adjustments [26, 28].

Safety of Diagnostic Ultrasound

Acoustic energy from US beams has direct interactions with the scanned tissues [20]. The physical effects on tissues can be both thermal and nonthermal. Although no direct causal relationship has been shown between the use of diagnostic ultrasound and adverse bioeffects in humans, the output of diagnostic ultrasound is regulated in the United States by the Food and Drug Administration (FDA) to minimize any potential bioeffects [21].

General Considerations

Minimizing the time that an ultrasound field is on is crucial in limiting patient exposure. By using periodic pulses, pulsed wave (PW) Doppler reduces the time under which a subject is irradiated when compared to continuous wave (CW) Doppler [21]. Spectral Doppler and M-mode imaging are stationary modes such that the exposure is along a single beam path. Imaging modes such as color flow imaging, power Doppler, and B-mode are scanned modes, and beams are primarily fired to different locations to increase the field of view such that the ultrasound power is more widely distributed.

Reduction of beam intensity, peak negative pressure, and exposure time are of prime importance in preventing unnecessary bioeffects [21]. Therefore, the as low as reasonably achievable (ALARA) principle is recommended when performing ultrasound examinations [29, 30]. Intensity can be measured as the temporal peak intensity, the greatest intensity at any point in time, or the spatial peak intensity, the largest intensity anywhere in the ultrasound field, for example, the focal point. The average intensity value during the ultrasound pulse is defined as the pulsed average intensity [20]. Another quantification of intensity is the temporal average intensity, which is the average value within the entire pulsed repetition period. For pulsed wave ultrasound, the temporal average intensity is related to the duty factor, or the fraction of time that the ultrasound beam is being transmitted through tissues [21]. A small duty factor signifies longer periods between pulses, decreasing the temporal average intensity. These intensity parameters serve as important factors when studying the possibility of bioeffects in tissues. Considering these values along with the overall exposure time, or dwell time, can increase operator awareness of the possibility of causing bioeffects [16–21]. The operators should follow the ALARA principle to guide how to perform ultrasound exams by using system controls to reduce ultrasound intensity and pressure, limit dwell time and examination time while maintaining image quality and efficient and effective diagnostic scanning and performing ultrasound exams only when medically indicated.

Thermal Effects

Acoustic energy is converted into heat within tissues. The risk of bioeffects, therefore, rises with increasing temperature and duration of exposure. The duration of exposure, type of tissue imaged, spatial focusing, rate of temperature increase, and ultrasound frequency can all influence the risk of bioeffects.

The thermal index (TI) can be used to assess potential heating from a transducer in specific equipment settings. It is displayed on ultrasound machines as a ratio of acoustic power currently being produced and the maximal acoustic power required to raise the tissue temperature 1°C in adults [16].

The American Institute of Ultrasound in Medicine (AIUM) provides a recommendation of the maximum scanning time at each TI where one can be reasonably assured against adverse bioeffects caused by thermal changes in patient tissue [30]. The TI may be displayed on US screens for operators to assess possible adverse effects based on the TI value and the duration of their imaging time. Monitoring the TI and reducing exposure time allows for risk reduction and minimization of temperature increases in keeping with the ALARA principle.

Greater temperature increases are possible in spectral Doppler, with less seen in color Doppler, and the least seen in B-mode imaging. But the TI is designed to help the operator minimize output regardless of the mode, while needing to keep in mind dwell time.

Acoustic Cavitation

Ultrasound interacts with gas bodies in a process known as cavitation, which may result in physical tissue changes [21]. Generated microbubbles can interact with the ultrasound to produce fluid motion, including jetting. Bubble collapse may lead to shearing forces within tissues and high temperatures in and around the bubbles. Cavitation can produce heat and free radicals, scatter acoustic energy, and microstreaming of fluid, all potential sources for mechanical changes [21].

Beam properties, such as frequency and pulse duration, have been shown to impact cavitation. Tissue fluid parameters, including density, viscosity, compressibility, heat conductivity, and surface tension, impact cavitation as well.

Ultrasound contrast agents contain microbubbles that have the potential for cavitation. Adverse effects of microbubbles have been demonstrated when operating with a mechanical index (MI) >0.4 in mammalian tissues when contrast agents are present. The mechanical index is defined in the following section. AIUM recommendations for limiting gas body bioeffects when operating at MI >0.4 include minimizing contrast dose, MI, and total exposure time [31].

Mechanical Index

As the TI serves to monitor the potential for thermal energy increases in tissues, the mechanical index (MI) is used to gauge the likelihood of mechanical effects due to cavitation [21]. MI is proportional to the acoustic pressure and inversely proportional to the square root of the ultrasound frequency. This means that higher acoustic pressures are needed for higher-frequency ultrasound to have the same potential for cavitation. The display of the MI allows the operator to more easily monitor this potential for effects. Studies have shown an MI of 0.4 as the threshold for pulmonary capillary hemorrhage and an MI of 1.4 for intestinal capillary hemorrhage. These values were reported for laboratory animals, and further research must be done to better understand possible bioeffects in humans [30, 32]. Again, the objective is to minimize the MI while maintaining diagnostic image quality.

Tissue Type

As mentioned earlier in this chapter, ultrasound beam absorption and attenuation vary with tissue type. Fluids such as amniotic fluid, blood, or urine have low ultrasound attenuation, while bone absorbs acoustic energy quite well. Reducing output power while increasing gain becomes important when imaging bony structures to minimize exposure, which is the reason the TI has three forms that are selectable on the ultrasound machines depending on the imaging situation. TIS is for soft tissue, TIB for bone distal to the transducer, and TIC for bone near the transducer, for example, transcranial ultrasound. More perfused tissues will better dissipate thermal energy and will be cooled down more effectively.

Current AIUM guidelines for obstetric ultrasound discuss that although no causal relationships have been shown between obstetric screening and harmful bioeffects, fetal imaging should only be done when medically indicated. Other nonmedical, or nonpurposeful, viewing of the fetus is contraindicated and inappropriate. When performing fetal imaging, the AIUM recommends that dwell times should be minimized, output indices such as the thermal and mechanical index monitored, and that Doppler and elasticity measurements are done appropriately and only when indicated [29, 33].

Soft tissues can shift acoustic energy to higher frequencies, causing amplitude distortion, which results in a shockwave that may expose tissues to disproportionally higher levels of energy [21]. For example, full bladder scanning can result in shockwave formation because fluid has a low attenuation coefficient and thus places higher ultrasound frequencies at posterior structures. Additionally, enhanced attenuation, caused by increasing the ultrasound frequency, can expose these structures to larger energy absorption.

Endocavitary uses of diagnostic ultrasound bring the transducer in closer proximity to sensitive tissue, and surface heating of the transducer itself may serve as a heat source. This increases the general concern for endocavitary scans. However, ultrasound, when used properly, is a safe and effective medical device whose benefits outweigh the potential risks.

Reduction of Ultrasound Output

Organizations, including the FDA and AIUM, helped develop the "Standard for Real-Time Display of Thermal and Mechanical Acoustical Output Indices on Diagnostic Ultrasound Equipment." This serves the purpose of displaying both the TI and MI on the US machine so operators are more aware of the acoustic beam interaction with tissues and will allow operators to perform risk-versus-benefit analyses when conducting ultrasound exams.

It is suggested that machines should be set to examination presets according to the type of examination performed, for example, an OB preset should be used for an OB examination to avoid unnecessary exposure. Direct controls, such as correct diagnostic settings, output power, and focusing, should be taken into consideration by operators to reduce US output. For example, the lowest output power that produces a clear image should be used as well as proper optimization of the beam focus at target tissues to avoid unnecessary exposure [21]. Indirect controls should also be considered. These affect the temporal and spatial distribution of the acoustic beam. Ultrasound imaging modes (B-mode, color Doppler) distribute ultrasound over a larger portion of tissue, while unscanned modes (continuous wave Doppler, spectral or pulsed Doppler, M-mode) apply the ultrasound along a single beam thus increasing the potential for heating. Pulsed repetition frequency (PRF) decreases can result in lower temporal average intensity and thus decrease potential heating. Reduction of burst length in pulses can also reduce patient exposure. Other factors include appropriate transducer selection and appropriate use of receiver gain controls. Image quality can be enhanced without increasing output intensity by using receiver gain and TGC [21].

Scanning time for training and research purposes should be kept to a minimum, and informed consent should be given after explaining the possibility of bioeffects and how the scan relates to what is commonly performed in a clinical examination.

For study purposes, ultrasound should be used conservatively to reduce repetitive and lengthy examinations [29]. Although ultrasound is safer compared to other diagnostic imaging such as CT, operators are responsible for informing patients of the overall safety, maximizing efficiency, and minimizing patient exposure to unnecessary ultrasound when performing any studies.

Knobology

There is considerable variation in the design of user controls between ultrasound vendors (Fig. 1.9). The operator needs to be familiar with the “knobology” (knobs, buttons, functions) on the machine being used to perform ultrasound examinations [34] and understand what these knobs control on the system. This is critical for obtaining high-quality clinical images in a safe and effective manner. For example, increasing the receive gain control or the power control can increase the echo brightness in a B-mode image. But the gain control does not increase the ultrasound exposure to the patient. Similarly, adjusting the TCG can improve image quality without increasing exposure. Thus, understanding how these controls can be used to make images more clinically useful, and which are changing the ultrasound exposure, is important knowledge and should be part of proper training. Most commercial ultrasound machines have functions to preset parameters of ultrasound image processing (image depth, total gain, dynamic range, harmonic imaging, speckle reduction) for specific ultrasound examination (e.g. cardiac, abdomen, OB, arterial, and venous) after selecting a transducer (e.g. linear array and phased array). Preset

FIGURE 1.9 Knobology of ultrasound machine.

functions potentially improve the performance of ultrasound examinations by using standardized, proper, and consistent image processing and quality.

Commonly used knobology is as follows [34]:

B (or 2D): B-mode (grayscale) ultrasound image
C (or CD): color flow imaging
CW: continue wave Doppler mode
D (P or PW): pulse wave Doppler mode
Depth: change image depth
G: total image gain
Freeze: freeze a frame of image
M: M-mode ultrasound
P (or store): save static image or real-time cine loop
TGC: time-gain compensation
Zoom: zoom in/out of image (or a region of interest)

Summary

Ultrasound principles and instrumentation have been refined for centuries and developed into a mainstay of medical diagnostic imaging. Choosing the correct type of transducer as well as using the appropriate imaging mode will create the optimal images necessary for any medical evaluation. Ultrasound operators must be able to identify physiological structures as well as artifacts for detecting pathologic changes while also minimizing the risks of ultrasound to the patients being scanned.

Review Questions

1. The propagation speed of a wave is determined by which of the following?
   a. Wavelength
   b. Amplitude
   c. Period
   d. Medium
   e. Frequency
2. Which of the following answer choices places media in order of increasing sound propagation speed?
   a. Gas, solid, liquid
   b. Gas, liquid, solid
   c. Solid, liquid, gas

- **d.** Liquid, solid, gas
- **e.** Solid, bone, liquid

3. If the wavelength of a medium at a given frequency is 4 mm and the frequency is doubled, what is the wavelength?

- **a.** 1 mm
- **b.** 2 mm
- **c.** 4 mm
- **d.** 16 mm
- **e.** 32 mm

4. Using given values of propagation speed and frequency, which of the following can be calculated?

- **a.** Period
- **b.** Wavelength
- **c.** Amplitude
- **d.** a and b
- **e.** a and c

5. If the frequency of sound is decreased by ½, how will it affect the wavelength and why?

- **a.** Stay the same
- **b.** Decrease, because frequency and wavelength are directly related
- **c.** Decrease, because frequency and wavelength are inversely related
- **d.** Increase, because frequency and wavelength are directly related
- **e.** Increase, because frequency and wavelength are inversely related

6. What transducer properties minimize the loss of lateral resolution of the far field?

- **a.** A larger transducer width
- **b.** Lower wavelength
- **c.** Higher frequencies
- **d.** Decreased scan line thickness
- **e.** All of the above

7. What is true for scanning deep tissues?

- **a.** High transmit frequency is used
- **b.** It is more difficult to maintain high frame rates
- **c.** Spatial resolution is maintained
- **d.** Low scan line densities are preferred
- **e.** Low attenuation in deep tissue

8. A sound beam traveling perpendicular to blood flow will demonstrate what kind of Doppler effect?

- **a.** Large Doppler shift
- **b.** Normal Doppler shift
- **c.** Small Doppler shift
- **d.** No Doppler shift
- **e.** The same Doppler shift as parallel to blood flow

9. Which two factors are located on the ultrasound device screen and should be monitored regularly by the operator to reduce chances of bioeffects?
   - **a.** Beam intensity, dwell time
   - **b.** Duty factor, thermal index
   - **c.** Thermal index, mechanical index
   - **d.** Pulsed average intensity, temporal average intensity
   - **e.** None of the above
10. Which strategy can be used to reduce ultrasound exposure and minimize bioeffects?
    - **a.** Selecting appropriate machine settings prior to starting subject screens
    - **b.** Avoiding repetitive and lengthy examinations
    - **c.** Using the lowest output intensity to produce a clear image
    - **d.** Move or lift the transducer when imaging of a stationary structure is not necessary
    - **e.** All the above

Correct answers: 1. d; 2. b; 3. b; 4. d; 5. e; 6. e; 7. b; 8. d; 9. c; 10. e.

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