Paula Ferrada
Editor

Ultrasonography in the ICU

Practical Applications

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This book is dedicated to residents and fellows who are learning the use of ultrasound to achieve better patient care. I truly believe we can affect patient outcome through education and innovation, and it is up to all of us learners to advance our field.

Preface

In the last decade ultrasound has become an extension of the physical exam. This is especially important when treating patients in extremis since it provides rapid information and does not require patient transport.

The use of this bedside tool has been made easier in order to bring critical care expertise to the location of the patient in need.

This volume illustrates practical applications of this tool, in an easy to understand, user-friendly approach. Because of its simple language and case-based teachings, this book is the ideal complement to clinical experience performing ultrasound in the critically ill patient.

Internet Access to Video Clip

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Contents

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1 Basics of Ultrasound

Irene W. Y. Ma, Rosaleen Chun and Andrew W. Kirkpatrick

Basics of Ultrasound

Ultrasound is increasingly used as a point-of-care device in the clinical arena, with applications in multiple clinical domains [1–6]. To be able to use ultrasound devices appropriately for its various applications, appropriate training, practice, and a requisite understanding of the basic physics of sound transmission are of paramount importance [7–14].

Generation of an ultrasound image relies on interpreting the effects of sound waves propagating in the form of a mechanical energy through a medium such as tissue, air, blood or bone. These waves are transmitted by the ultrasound transducer as a series of pulses, alternating between high and low pressures, transmitted over time (Fig. 1.1a, b). As they are transmitted, these sound waves mechanically displace molecules locally from their equilibrium. Compression occurs during pulses of high pressure waves, causing molecules to be pushed closer together, resulting in a region of higher density (see Fig. 1.1a), while rarefaction occurs during pulses of low pressure waves, causing molecules to be farther apart and less dense. Once transmitted, these sound waves interact within tissue. Based on the select properties of the sound waves transmitted as well as properties of the tissue interfaces, some of these sound waves are then reflected back to the transducer, which also acts as a receiver. The signals are then processed and displayed on the monitor as a two-dimensional (2-D) image. This type of image is the typical image used in point-of-care imaging and is known as B-mode (or brightness mode) for historical reasons.

Frequency, Period, Wavelength, Amplitude, and Power

A number of parameters are used to describe sound waves, and some of these have direct clinical relevance to the user. These parameters include frequency, period, wavelength, amplitude, and power.

Frequency is the number of waves passing per second, measured in hertz (Hz). Two closely related concepts are the *period* (p), which is the time required for one complete wave to pass, measured in microseconds (μs) and *wavelength* (λ), which is the distance travelled by one complete wave, measured in millimeters (mm) (see Fig. 1.1a). Frequency is inversely related to period and wavelength. That is, the shorter the

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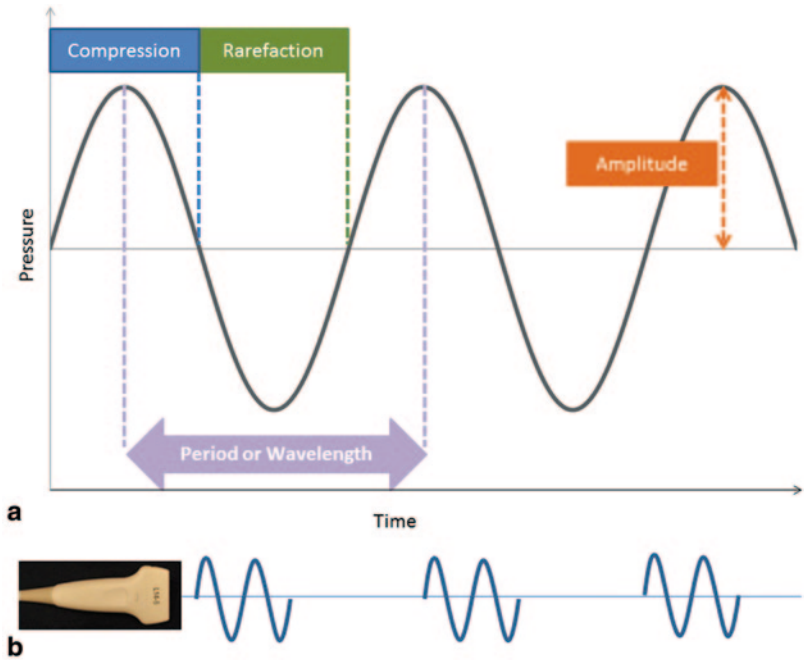

Fig. 1.1 **a** Sound waves transmitted propagating through a medium, alternating between high and low pressures, transmitted over time. Compression occurs during high pressure waves, pushing molecules mechanically closer together. Rarefaction occurs during low pressure waves, causing molecules to be farther part. Period refers to the time required for one sound wave to pass. Wavelength refers to the distance travelled by one complete sound wave. Amplitude refers to the height of the wave. **b** Transmission of a series of pulses of sound waves by a transducer

period, the higher the frequency; the shorter the wavelength, the higher the frequency. Ultrasound equipment typically operates within the range of 1 megahertz (MHz) to 20 MHz, which is well above the range of human hearing, generally considered to be between 20 to 20,000 Hz (0.00002 to 0.02 MHz). An understanding of frequency is clinically relevant to the operator and users of ultrasound. Specifically, choosing an appropriate frequency range will affect both the resolution of the image as well as the ability to penetrate tissues and image structures at the desired depth.

Frequency is one of the factors determining spatial resolution. Spatial resolution refers to the ability of ultrasound to distinguish between two objects in close proximity to one another as being distinct objects. Higher frequency sound waves yield better resolution than lower frequency waves. However, this improved resolution for higher frequency sound waves is at the expense of lower penetration [15]. That is, higher frequency sound waves are less able to image structures that lie further away from the transducer than lower frequency sound waves. Therefore, for typical applications in the intensive care unit, higher frequencies are more useful for imaging superficial structures while lower frequencies are more useful for imaging deeper structures. Thus, transducers with frequency ranges of 5 to 15 MHz are used for imaging superficial structures such as superficial vascular anatomy while ranges of 2 to 5 MHz are used for imaging deeper structures such as intra-abdominal organs.

Amplitude refers to the strength of the sound wave, as represented by the height of the wave (see Fig. 1.1a). Amplitude is measured in units of pressure, Mega Pascals (MPa). *Power* of the sound wave, refers to the total amount of energy in the ultrasound beam, and is measured in watts [16]. Power and amplitude are closely related, with power being proportional to the square of the amplitude [17]. In using ultrasound, one must keep in mind that for instance, by only doubling the amplitude, four times the energy is being delivered to the patient.

Understanding concepts regarding amplitude and power is critical to appreciate in facilitating the safe use of ultrasound. In general, the performance of ultrasound scans should comply with the ALARA (as low as reasonably achievable) principle by keeping total ultrasound exposure as low as reasonably achievable [18]. All ultrasound machines capable of exceeding a pre-specified output are required to display two output indices on the output display: Mechanical Index (MI), which provides an indication of risk of harm from mechanical mechanisms, and Thermal Index (TI), which provides an indication of risk of harm from thermal effects [18, 19]. The higher the indices, the greater the potential for harm. The Food and Drug Administration (FDA) regulations allow a global maximum MI of ≤1.9, except for ophthalmic applications, where the maximum allowed TI should be ≤1.0 and MI ≤0.23 [20]. For obstetrical applications, the current recommendations are for MI and TI to be ≤1.0 and the exposure time to be as short as possible: generally 5 to 10 min and not exceeding 60 min [21, 22].

Generation of Sound Waves

The generation of sound waves was made possible by the discovery of the piezoelectric effect in 1880: certain crystals vibrate when a voltage is applied to it, and conversely, subjecting the crystal to mechanical stress will result in an electrical charge [23]. Utilizing this principle, the transducer of an ultrasound machine houses crystal elements (Fig. 1.2), such that by applying electrical energy through the cable to these piezoelectric crystals, they change shape, vibrate, and in so doing, convert electrical energy into mechanical energy. Conversely, the piezoelectric crystals can also convert mechanical energy back into electrical energy, thereby allowing it to act as both a transmitter and a receiver. Within the transducer, the piezoelectric crystal is supported by the backing material (see Fig. 1.2), which serves to dampen any backward-directed vibrations, while the lens in front of the crystal serves to assist with focus. Finally, the impedance matching layer in front of both the piezoelectric elements and the lens assists with the transmission of sound waves into the patient [24]. Together, these components allow the transmission and receiving of sound waves. Irrespective of the characteristics of the transmitted sound waves, all ultrasound imaging relies on users interpreting the display of sounds waves reflected back to the receiver. Thus, an understanding of how sound waves travel and reflect from tissue is critical knowledge for any sonographer.

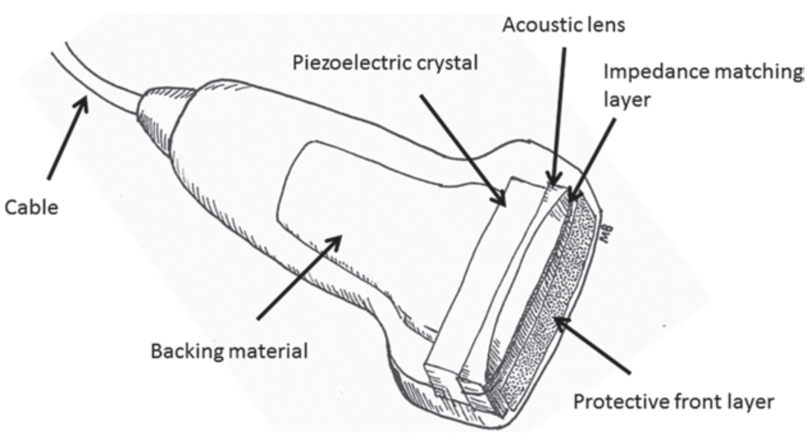

Fig. 1.2 A schematic representation of components of an ultrasound transducer. Illustration Courtesy of Mary E. Brindle, MD, MPH

Interactions of Sound Waves with Tissue

In order to understand how an ultrasound image is generated, it is important to understand the many ways in which sound waves propagate through and interact with tissue. Tissue characteristics such as density, stiffness, and smoothness, and surface size of the object being interrogated, all play critical roles in determining the amount of signal reflected back to the transducer. As only sound waves reflected back can assist in generating an image, it is critically important for the users to recognize how sound waves return to the transducer as well as how they fail to do so.

Propagation Velocity

The speed at which sound waves propagate within tissue is measured in meters per second (m/s). This velocity is determined by the density and stiffness of the tissue, rather than by characteristics of the sound waves themselves. Propagation velocity is inversely proportional to tissue density and directly proportional to stiffness of the tissue [17]. In other words, the denser the tissue, the slower the propagation velocity through that tissue, while the stiffer the tissue, the higher the velocity. In general, propagation speed is slowest through air (330 m/s) and fat (1450 m/s) and fastest through muscle (1580 m/s) and bone (4080 m/s) (Table 1.1) [25]. The average velocity through soft tissue is 1540 m/s, and it is this velocity that the ultrasound machine assumes its sound waves are travelling, irrespective of whether or not that is the case.

Understanding propagation velocities of different tissues is important for three reasons. First, propagation velocities through different tissue interfaces determine the amount of sound wave reflections, which in turn, determines the brightness of the signal display. Second, differences in propagation velocities are an important source of artifacts (see the section "Speed Propagation Error"). If the sound waves travel through tissue at a slower velocity than is assumed by the machine (e.g., through air or fat), any wave reflections from the object of interest will be placed at a farther distance on the display from the transducer than the true distance. Finally, as all diagnostic ultrasound uses the above mentioned approximation of ideal tissue characteristics, ultrasound will never yield the same fidelity of imaging as computer tomography (CT) or magnetic resonance imaging (MRI).

When sound waves interact with tissue, any or all the following processes may occur: reflection, scattering, refraction, absorption, and attenuation [15].

Reflection

When ultrasound waves propagate through tissue and encounter interfaces between two types of tissue, some of the sound waves will be reflected back. This reflected sound wave is called an *echo*. As previously mentioned, ultrasound imaging hinges upon the production and detection of these reflected echoes. Production of an echo is critically dependent upon the presence of an *acoustic*

Table 1.1 Propagation velocity in various media, measured in meters per second [25]. Acoustic impedance, measured in kilogram per meter squared per second [62, 63]. Attenuation coefficient, measured in dB/cm/MHz [25]

Medium	Propagation velocity (meters/second)	Acoustic impedance (kg/(m^2s))	Attenuation coefficient (dB/cm/MHz)
Air	330	430	10.00
Fat	1450	1.33×10^6	0.63
Water	1480	1.48×10^6	0.00
Average soft tissue	1540		0.70
Liver	1550	1.66×10^6	0.94
Kidney	1560	1.64×10^6	1.00
Blood	1570	1.67×10^6	0.18
Muscle	1580	1.71×10^6	1.30 (parallel)—3.30 (transverse)
Bone	4080	6.47×10^6	5.00

impedance difference between the two tissue types. Acoustic impedance is a property of the tissue, and is defined as the product of its tissue density and the propagation velocity of sound waves through that tissue. If two tissue types have identical acoustic impedance, then no echo will be produced, as no sound waves will be reflected back.

The brightness of the signal is directly related to the amount of reflection, and that the amount of reflection is proportional to the absolute difference in acoustic impedance between the two media. It therefore follows that a large acoustic impedance mismatch between two tissue types will result in a bright echogenic signal, while a small acoustic impedance mismatch between another two tissue types will result in an echo-poor signal. For example, at the interface between the liver and kidney, because of a minimal acoustic impedance difference between the two tissues, only about 1 % of the sound is reflected (see Table 1.1). Thus the interface between the kidney and the liver is somewhat harder to distinguish from one another (Fig. 1.3a) and less echogenic than the interface between muscle and bone, which has a large

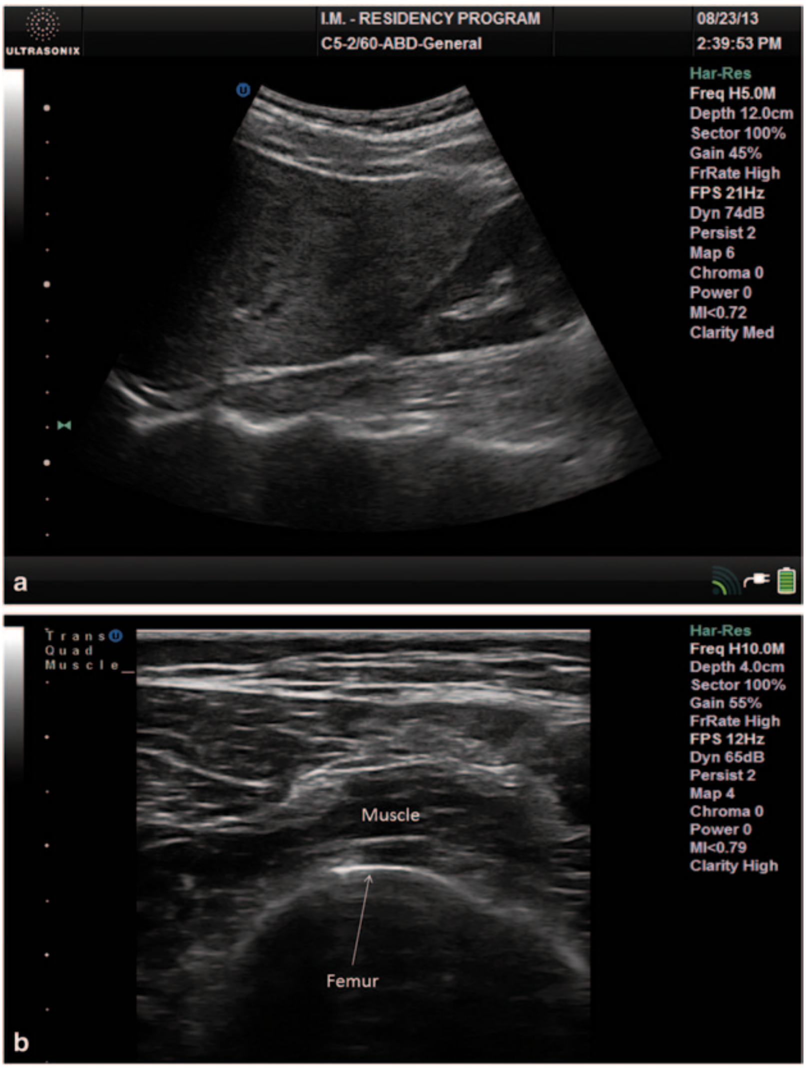

Fig. 1.3 **a** A longitudinal, oblique ultrasound view of liver and right kidney. Small acoustic impedance difference between liver and kidney results in a minimally echogenic interface between the two organs. **b** A transverse ultrasound view of the quadriceps muscle. Large acoustic impedance difference muscle and femur results in a bright echogenic interface between the two structures

acoustic impedance mismatch, resulting a bright echogenic line (see Fig. 1.3b). Finally, because of the very large acoustic impedance difference between tissue and air, upon encountering air, >99.9% of the sound waves are reflected. This results in minimal further propagation of sound waves. Therefore, beyond that interface, there is limited to no ability to further directly image structures [24]. This large acoustic impedance difference between air and skin is also the reason why coupling gel must be used for imaging purposes. Application of gel eliminates any air present between the transducer and the skin, assisting in the transmission of sound waves, rather than having most of them reflected back.

A second factor that determines the amount of reflection is the smoothness of the surface. For smooth surfaces that are large, compared with the size of the ultrasound's wavelength, *specular reflection* occurs (Fig. 1.4), resulting in a robust amount of reflection. However, for surfaces that are rough, where the undulations of the surfaces are of a similar size to the size of the ultrasound's wavelength, sound waves are reflected in multiple directions. This results in *diffuse reflection* (Fig. 1.5) [26]. Because the returning echoes are in multiple directions, only a few of them are received back on the transducer. As a result, diffuse reflection results in a less echogenic signal.

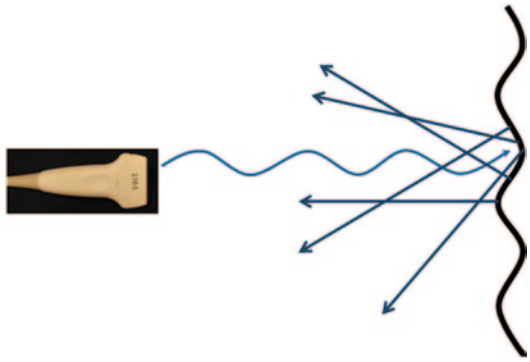

Fig. 1.5 Diffuse reflection occurs when sound waves are reflected off a rough surface of a similar size to the size of the wavelength

Scattering and Refraction

Additional ways in which emitted ultrasound waves do not reflect fully back to the transducer, resulting in attenuation of sound waves include scattering and refraction. Scattering occurs when ultrasound waves encounter objects that are small compared to the size of the ultrasound's wavelength, [15] which serves to diminish the intensity of the returned signal (Fig. 1.6).

Refraction occurs when sound waves pass from one medium to another with differing propagation velocities. These differing velocities

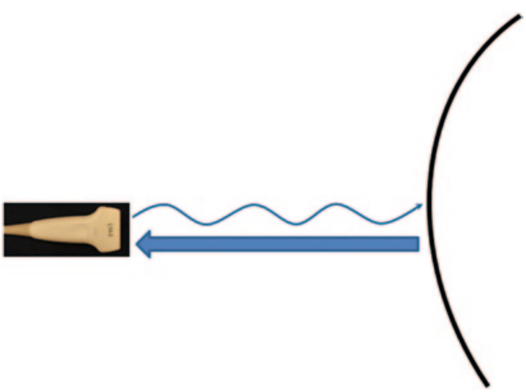

Fig. 1.4 Specular reflection occurs when sound waves are reflected off a smooth surface that is large compared with the size of the wavelength

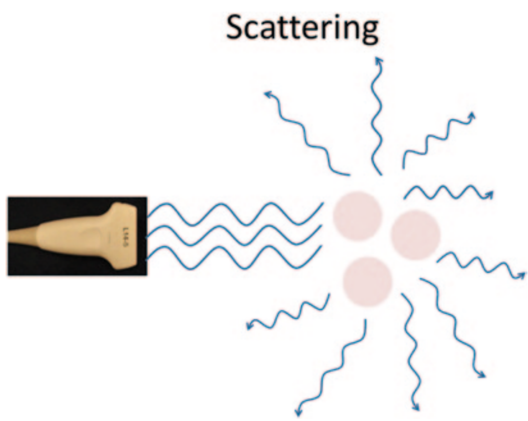

Fig. 1.6 Scattering occurs when sound waves are reflected off objects that are small compared with the size of the wavelength

result in refraction, or change in the direction of the original (or incident) sound wave [25]. The refracted angle, or magnitude of the change in direction of the ultrasound wave, is determined by Snell's law using the following equation:

$$\sin \theta_1 / V_1 = \sin \theta_2 / V_2$$

where θ_1 is the angle of incidence in the first medium, V_1 is the propagation velocity of sound in the first medium, θ_2 is the angle of refraction, and V_2 is the propagation velocity of sound in the second medium (Fig. 1.7). As can be seen from the equation, the higher the difference between the propagation velocities in the two media, the larger the magnitude of angle change of the refracted beam. Because the ultrasound machine assumes that the sound wave travels in a straight line and does not know that the sound path has been altered by refraction, [24] this results in artifacts such as the double-image artifact (see the section "Refraction Artifacts"). Thus, to minimize refraction, except for Doppler applications (see the section "The Doppler Effect"), an ultrasound image should be obtained at an angle as perpendicular as possible to structure of interest, in order to minimize the angle of incidence (Fig. 1.8a, b).

Absorption and Attenuation

As sound waves propagate through tissue, part of the acoustic energy is absorbed and converted into heat. The amount of absorption that occurs is a function of the (1) sound wave frequency, (2) scanning depth, and (3) the nature of the tissue itself.

Higher frequency sound waves are absorbed more than lower frequency sound waves. As stated earlier in this chapter, although higher frequency sound waves yield better resolution than lower frequency sound waves, this improved resolution is gained at the expense of lower penetration [15]. The inability of high frequency sound waves to penetrate deeply into tissue is a direct result of high absorption and conversion of acoustic energy into heat. Thus, a shallower depth, provided it captures sufficiently the structure of interest in the field of view, will result in

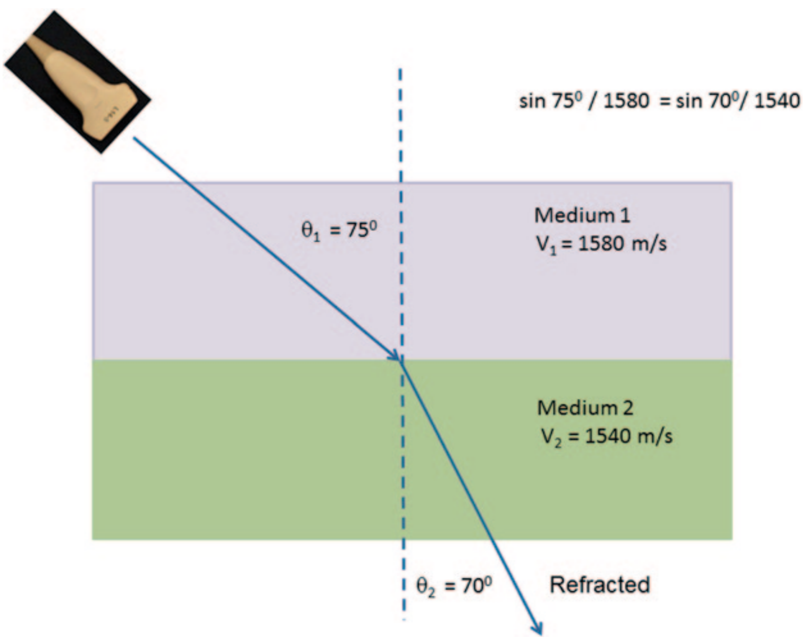

Fig. 1.7 Refraction occurs when sound waves pass from one medium with a propagation velocity to another medium with a differing propagation velocity