TEXTBOOK OF
DIAGNOSTIC ULTRASONOGRAPHY

SANDRA L. HAGEN-ANSERT,

SECOND EDITION

TEXTBOOK OF

DIAGNOSTIC ULTRASONOGRAPHY

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Foreword

It is a distinct pleasure for me to write the foreword for this important textbook of diagnostic ultrasonography. Since the introduction of the first edition in 1978, it has become widely acclaimed by professionals in the field for the high quality of its images and the clarity of the accompanying text. It is only natural that a second edition should be produced to incorporate the rapid changes in technology that have occurred in the interval.

The second edition contains important new sections on pediatric echocardiography, breast sonography, and the evolving field of superficial organ scanning. The physics and biologic effects sections have been dramatically revised to better meet the needs of practicing ultrasonographers.

I believe that I am in a unique position to preface this text since it was my good fortune to introduce the author to the field 13 years ago. The science or art of ultrasonography in those days was quite limited when compared to modern practice. Nevertheless, it was always exciting, and each new day seemed to bring forth an interesting finding that would at least temporarily perplex us. The close collaboration that existed between us has become typical of this important field. There is no question that today's ultrasonographers command considerable respect for their dedication and advancement of the profession. Sandra Hagen-Ansert is directly responsible for much of this development, since upon leaving our laboratory she has worked tirelessly in educating both physicians and ultrasonographers.

I am sure that she has learned (as have I) that the greatest professional joy results from stimulating others to ever increasing achievement and better patient care. It is to that goal that this text is directed.

George R. Leopold, M.D.

Preface
TO THE SECOND EDITION

At the completion of this revised edition I no longer wonder why there are not more complete textbooks on diagnostic ultrasonography available. The task of compiling all the material necessary to complete such a feat is enormous and one that could not be done without the cooperation of excellent contributors, enthusiastic students, and a supportive departmental staff.

The primary goal of preparing such a textbook was to have information on all areas of sonography available in one source textbook. A number of areas have been updated to include real time and automation techniques as they apply to current diagnostic practice. New chapters have been added to cover applications to the breast, neonates, and superficial structures.

Since there are so many excellent atlas textbooks available on ultrasonography, emphasis is placed on information that the sonographer needs to know and understand in order to be a quality diagnostic ultrasonographer. In an effort to help the sonographer understand the total clinical picture that the patient presents prior to the sonographic examination, anatomy, physiology, laboratory data, clinical signs and symptoms, pathology, and sonographic findings are found in each specific chapter.

I would like to extend my sincere appreciation to the following individuals who helped to make this edition possible: William Zwiebel, M.D., for his help on the liver chapter and in finding interesting cases; Chris Labinski, Rhonda Aborgast, and Jackie Cassidy for their continued support; Barbara VanderWerff, Becky Levzow, and Susan Yourd, for their comments and critiques; Tom Yourke, Jeannie McFadden, Lorrie Stadtmueller, Judy McClellan, John Mayer, Marty Gebhart, and Earl Bell, for their review of the manuscript; Robert Vennie, for his photographic assistance; Jeffrey Allyn Slade, for his medical illustrations; Shirley Wikum and Cindy Johnson, for their secretarial assistance; Jeff Brown, M.D., and Charlie Austin, M.D., for their encouragement and support; Tom Lawson, M.D., and Vicki Vieaux, for their Octoson images from Milwaukee County Medical Center; Marcia Lavery, for her real time images and protocol from the New England Deaconess Hospital; Jean Corneil, for her photographic support to portions of the echocardiography chapters; Harry Rakowski, M.D., and Bob Howard, M.D., for their support and contributions to chapters in adult and congenital heart disease; and Don Ladig, Rosa Kasper, Karen Edwards, and especially George Stericker at C.V. Mosby, for their endless hours and support in completing this textbook.

Sandra L. Hagen-Ansert

Preface
TO THE FIRST EDITION

Medicine has always been a fascinating field. I was introduced to it by Dr. Charles Henkelmann, who provided me with the opportunity to learn radiography. Although x-ray technology was interesting, it was not challenging enough. It did not provide the opportunity to evaluate patient history or to follow through interesting cases, which seemed to be the most intriguing aspect of medicine and my primary concern.

Shortly after I finished my training, I was assigned to the radiation therapy department, where I was introduced to a very quiet and young, dedicated radiologist, whom I would later grow to admire and respect as one of the foremost authorities in diagnostic ultrasound. Convincing George Leopold that he needed another hand to assist him was difficult in the beginning, and it was through the efforts of his resident, Dan MacDonald, that I was able to learn what has eventually developed into a most challenging and exciting new medical modality.

Utilizing high-frequency sound waves, diagnostic ultrasound provides a unique method for visualization of soft tissue anatomic structures. The challenge of identifying such structures and correlating the results with clinical symptoms and patient data offered an ongoing challenge to the sonographer. The state of the art demands expertise in scanning techniques and maneuvers to demonstrate the internal structures; without quality scans, no diagnostic information can be rendered to the physician.

Out initial experience in ultrasound took us through the era of A-mode techniques, identifying aortic aneurysms through pulsatile reflections, trying to separate splenic reflections from upper-pole left renal masses, and, in general, trying to echo every patient with a probable abdominal or pelvic mass. Of course, the one-dimensional A-mode techniques were difficult for me to conceptualize, let alone believe in. However, with repeated success and experience from mistakes, I began to believe in this method. The conviction that Dr. Leopold had about this technique was a strong indicator of its success in our laboratory.

It was when Picker brought our first two-dimensional ultrasound unit to the laboratory that the "skeptics" started to believe a little more in this modality. I must admit that those early images were weather maps to me for a number of months. The repeated times I asked, "What is that?" were enough to try anyone's patience.

I can recall when Siemens installed our real-time unit and we saw our first obstetric case. Such a thrill for us to see the fetus move, wave its hand, and show us fetal heart pulsations.

By this time we were scouting the clinics and various departments in the hospital for interesting cases to scan. With our success rate surpassing our failures, the case load increased so that soon we were involved in all aspects of ultrasound. There was not enough material or reprints for us to read to see the new developments. It was for this reason that excitement in clinical research soared, attracting young physicians throughout the country to develop techniques in diagnostic ultrasound.

Because Dr. Leopold was so intensely interested in ultrasound, it became the diagnostic method of choice for our patients. It was not long before conferences were incomplete without the mention of the technique. Later, local medical meetings and eventually national meetings grew to include discussion of this new modality. A number of visitors were attracted to our laboratory to learn the technique, and thus we became swamped with a continual flow of new physicians, some eager to work with ultrasound and others skeptical at first but believers in the end.

Education progressed slowly at first, with many laboratories offering a one-to-one teaching experience. Commercial companies thought the only way to push the field was to develop their own national training programs, and thus several of the leading manufacturers were the first to put a dedicated effort into the development of ultrasound.

It was through the combined efforts of our laboratory and commercial interests that I became interested in furthering ultrasound education. Seminars, weekly sessions, local and national meetings, and consultations became a vital part of the growth of ultrasound.

Thus, as ultrasound grew in popularity, more intensified training was desperately needed to maintain its initial quality that its pioneers strived for.

Through working with one of the commercial ultrasound companies conducting national short-term training programs, I became acquainted with Barry Goldberg and his enthusiasm for quality education in ultrasound. His organizational efforts and pioneer spirit led me to the east coast to further develop more intensive educational programs in ultrasound.

Through these experiences the need for a diverse ultrasound textbook was shown. Thus this text was written for the sonographer involved in clinical ultrasound, with emphasis on anatomy, physiology, pathology, and ultrasonic techniques and patterns. Clinical medicine and patient evaluation are important parts of the ultrasonic examination and as such are discussed as relevant to pathology demonstrated by ultrasound.

It is my hope that this textbook will not only introduce the reader to the field of ultrasound but also go a step beyond to what I have found to be a very stimulating and challenging experience in diagnostic patient care.

I would like to acknowledge the individual who contributed most to my early interest in diagnostic ultrasound, George R. Leopold, M.D., for his personal perseverance and instruction, as well as for his outstanding clinical research. My thanks also to Dr. Sam Halpern for the encouragement to publish; to Dr. Barry Goldberg for the opportunity to develop training programs in an independent fashion and for his encouragement to stay with it; to Drs. Barbara Gosink, Robert O'Rourke, Mike Crawford, and David Sahn for their encouragement throughout the years at U.C.S.D.; to Drs. Jagdish Patel and Carl Rubin for their continued interest in developing ultrasonic techniques; to Dr. Daniel Yellon for his early hours of anatomy dissection and instruction in clinical cardiology; to Dr. Carson Schneck for his excellent instruction in gross anatomy and sections of "Geraldine"; to Dr. Harvey Watts for his help in the preparation of the gross anatomy pathology photographs from Episcopal Hospital; to Dr. Jacob Zatuchni for the interest, enthusiasm, and understanding he showed me while at Episcopal Hospital; to Drs. Paul Walinski and Edward Säcks for their enthusiastic support in echocardiology; to Reuben Mezrich, David Vilkomberson, Ray Wood, Joe Geck, and Nate Pinkney for their continued support and participation in the physics chapter; to Marcia Lavery for her support with the liver chapter; to John Dietz for the photography of the equipment and patient positions; to Bill Burke, medical illustrator, for his aid in the preparation of the photographs and cardiac illustrations; to Arthur J. Ansert, Jr., who provided the atmosphere of productivity to complete such a book.

The students in diagnostic ultrasound from Episcopal Hospital and Thomas Jefferson University Medical Center continually work toward the development of finer ultrasound techniques and instruction, and for their support I would like to thank them.

A special acknowledgment is made to the many contributors of various chapters within the textbook. Much of this information was accumulated as part of their student participation in the Ultrasound Program at Episcopal Hospital and Thomas Jefferson University Medical Center.

Sandra L. Hagen-Ansert

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Physics of diagnostic ultrasound

JAMES A. ZAGZEBSKI

NATURE OF SOUND WAVES

The passage of sound through a medium involves wave propagation, in which particles within the medium are caused to vibrate about their *rest* position. This disturbance propagates through the medium at a speed determined by the properties of the medium itself. Passage of the wave results in the transfer of *energy* through the medium. However, there is no net transfer of particles (i.e., after a sound wave has passed through the medium the particles return to their normal, equilibrium position; we are assuming that the strength of the wave is low enough to allow this latter statement to be made). Sound waves can be transmitted through many materials, such as air, water, wood, plastic, and biologic tissues. They cannot be transmitted through a vacuum because they require some form of matter for their propagation.

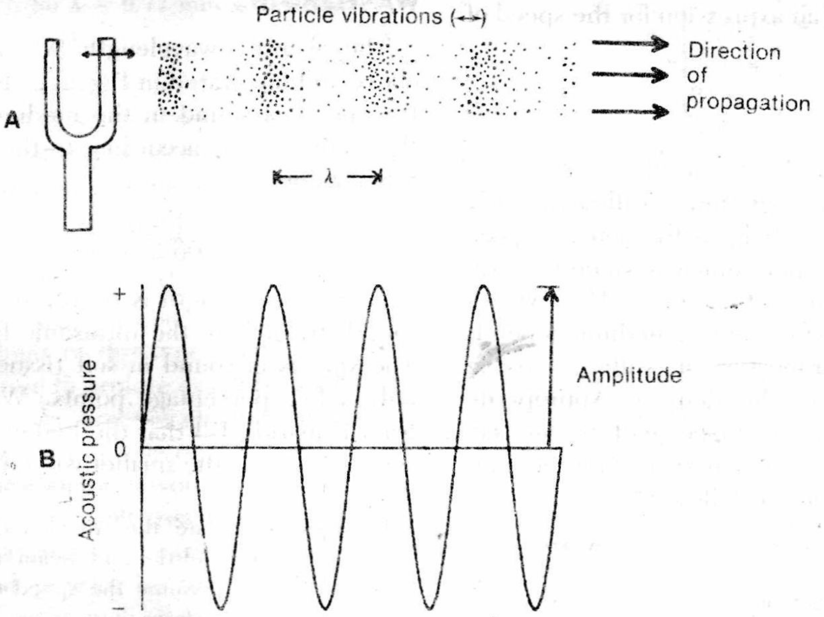

Fig. 1-1. Generation of a sound wave in air by a vibrating tuning fork. The graph shows the acoustic pressure versus distance at some instant of time.

Sound waves are produced by vibrating sources. One of the simplest examples of a source of sound is a tuning fork vibrating in air (Fig. 1-1, *A*). The vibrations of the tuning fork cause adjacent molecules in the air to be compressed together and drawn apart, depending on the direction of movement of the arm of the tuning fork. Molecules that are compressed together push other molecules closer together, which push other farther molecules closer together, etc.; thus the acoustic disturbance propagates outward.

A tuning fork vibrates back and forth in a regular fashion, sometimes referred to as *simple harmonic motion*. The resultant air compressions are accompanied by increases in the pressure. If it were possible to measure the pressure at different points near the tuning fork at any instant of time, the measurement results would appear as in Fig. 1-1, *B*. The pressure varies with distance, tracing out a *sine wave*, as shown. Here *0* pressure refers to equilibrium, ambient conditions, usually the atmospheric pressure if we are considering a sound wave in air. Places where particles are squeezed together are referred to as regions of *compression* and the pressure here is greater than 0. The maximum pressure swing occurring during passage of the wave is called the *pressure amplitude*, also defined in the figure. Places where the particles are drawn apart are referred to as regions of *rarefaction* and the pressure here is less than 0. The distance over which the curve repeats itself is called the acoustic *wavelength*, given by the symbol λ in the figure.

Just as the vibrating tuning fork does not remain stationary, so a plot of pressure versus distance also varies from one instant to the next (Fig. 1-2). This is because the sound wave is propagating outward from the source. A useful way of expressing the temporal behavior of a sound wave is to plot the pressure versus time at a single point in the medium. The resultant curve also traces out a *sine wave* (Fig. 1-3). The number of times per second the disturbance is repeated at any point is called the *frequency*. The time it takes for the disturbance to repeat itself is the *period*, labeled T in Fig. 1-3. Frequency, f, and period, T, are inversely related; that is,

$$T = \frac{1}{f} \tag{1-1}$$

Example: Suppose the period of a waveform is 0.5 second. Calculate the frequency.

Solution: You can rearrange Equation 1-1 by multiplying both sides of the equation by f and dividing both sides by T. The result is

$$f = \frac{1}{T}$$

Substituting gives

$$f = \frac{1}{0.5 \text{ sec}} = 2/\text{sec}$$

In other words, if the period is 0.5 second, the frequency is 2 times per second.

Fig. 1-4 shows that as the period decreases, the frequency increases, and vice versa.

TYPES OF SOUND WAVES

Sound waves are mechanical vibrations that propagate in a medium. In response to the sound wave, particles in the medium are displaced from their rest position and vibrate back and forth. In the example in Fig. 1-1 the particle displacement is in the same direction as the wave propagates. This mode of vibration is referred to as *longitudinal wave* propagation. Other types of vibrations are possible, depending on the type of medium. For example, transverse vibrations or shear waves may be transmitted through solid materials. These are characterized by particle vibrations perpendicular to the direction of vibration (Fig. 1-5). In this textbook we are concerned mainly with propagation of sound in the soft tissues of the body. Only longitudinal waves are of interest here because this is the only mode of vibration that can be transmitted through soft tissue.

FREQUENCY

It was mentioned earlier that the sound frequency is the number of oscillations per second that the source or the particles in the medium make as they vibrate about their rest position. The unit for frequency is *cycles per second* or *hertz*. Commonly used multiples of 1 hertz are as follows:

1 cycle per second = 1 hertz = 1 Hz

1000 cycles per second = 1000 hertz = 1 kilohertz = 1 kHz

1,000,000 cycles per second = 1,000,000 hertz = 1 megahertz = 1 MHz

The metric notation will be used consistently in this book. Appendix F gives the more common metric prefixes and their decimal equivalents.

A classification scheme for acoustic waves according to their frequency is given in Fig. 1-6. Most humans can hear sound if it has a frequency in the range of 15 Hz to approximately 15 to 18 kHz. This is referred to as the *audible frequency* range. Frequencies greater than 20 kHz are referred to as *ultrasonic*. Vibrations whose frequencies are below the audible range are termed *infrasonic*. Examples of infrasonic transmissions include vibrations introduced by air ducts, ocean waves, and seismic waves.

The ultrasonic frequency range is used extensively, both by humans and by animals. Except for therapy ultrasound, most medical applications utilize frequencies that lie in the 1-to-20-MHz range.

SPEED OF SOUND

The speed with which acoustic waves propagate through a medium is determined by the *characteristics of the medium* itself. (There are slight dependences on other factors, such as the ultrasonic frequency, but these are so small that they can be ignored completely in our discussion.) Specifically, for longitudinal sound waves in either liquids or body tissues an expression for the speed of sound, c, is

$$c = \sqrt{\frac{B}{\rho}} \tag{1-2}$$

In this equation B refers to the elastic properties of the medium and is called the *bulk modulus*. The symbol ρ is the density, given in g/cm^3 (grams per cubic centimeter) or kg/m^3 (kilograms per cubic meter). Thus we see that the speed of sound in a medium depends on the elastic properties, or "stiffness," of the medium and on the density. Appropriate units for speed are m/sec (meters per second). The speeds of sound in some nonbiologic materials are as follows[1,6]:

	m/sec
Air	330
Silastic materials	950
Ethyl alcohol	1177
Water	1480
Lead	2400
Crown glass	6120
Aluminum	6400

The speed of sound in biologic tissues is an important parameter in imaging applications.[4] Values that have been measured in different human tissues are as follows[6]:

	m/sec
Lung	600
Fat	1460
Aqueous humor	1510
Liver	1555
Blood	1560
Kidney	1565
Muscle	1600
Lens	1620
Skull bone	4080

The lowest speed shown is that for lung tissue, due to the presence of air-filled alveoli in this tissue. Most tissues of concern to us, that is, those through which sound can be readily propagated in the megahertz frequency range, have speeds of sound in the neighborhood of 1500 to 1600 m/sec. Fat is seen to come out on the low end of this chart whereas muscle tissue and the lens of the eye come out on the high-speed end. Measurements of the speed of sound in bone tissue result in values two to three times those recorded in most soft tissues.

The average speed of sound in soft tissues (excluding the lung) is 1540 m/sec, and range-measuring circuits on many diagnostic ultrasound instruments are calibrated on this basis. Close inspection of the biologic tissue list above reveals that the propagation speed in every soft tissue of concern to us in diagnostic ultrasound is within a few percentage points of 1540 m/sec.

WAVELENGTH

The acoustic wavelength (λ), as defined above and illustrated in Fig. 1-1, depends on the speed of sound in the medium, c, and the frequency, f, according to the following relationship:

$$\lambda = \frac{c}{f} \tag{1-3}$$

Thus the wavelength is simply the speed of sound divided by the ultrasonic frequency. The speeds of sound in soft tissues vary by only a few percentage points. We can see from Equation 1-3 that the higher the ultrasonic frequency the smaller will be the wavelength.

Example: Calculate the wavelength for a 2-MHz ultrasound beam in soft tissue. Assume the speed of sound is 1540 m/sec.

Solution: The wavelength can be calculated directly using Equation 1-3, with c = 1540 m/sec and f = 2 MHz = 2 × 10^6 cycles/sec.

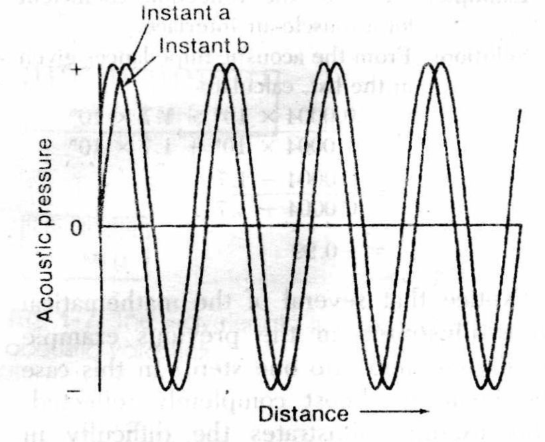

Fig. 1-2. Acoustic pressure versus distance at two different times. Same setup as in Fig. 1-1. The two curves are identical except for being slightly out of phase.

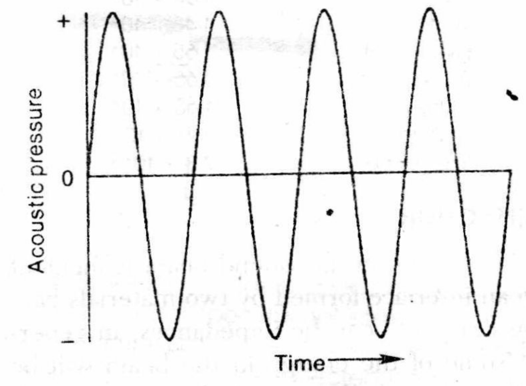

Fig. 1-3. Pressure versus time measured at a single point.

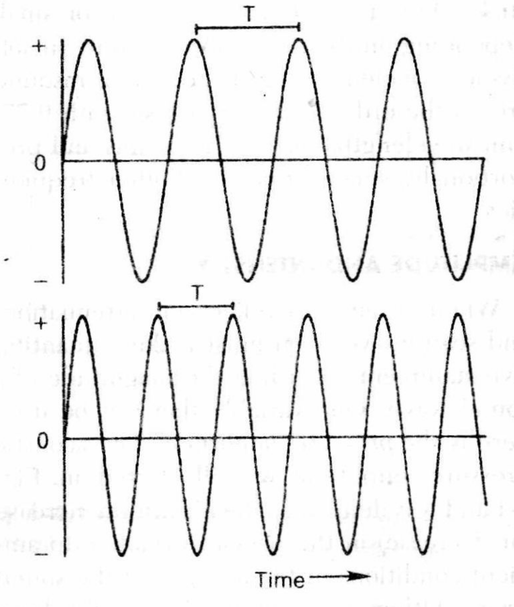

Fig. 1-4. Sine waves of two different frequencies.

$$\text{Thus } \lambda = \frac{1540 \text{ m/sec}}{2 \times 10^6 \text{ cycles/sec}}$$
$$= 0.0077 \text{ m/cycle}$$
$$= 0.77 \text{ mm/cycle}$$

We always drop the */cycle* since it is obviously included in our designation *wavelength*.

So $\lambda = 0.77$ mm is the correct answer.

You may wish to study the material in Appendix F at this stage to review metric conversions. Appendix E also contains examples of addition, subtraction, multiplication, and division in which numbers are expressed as exponentials (i.e., 2,000,000 cycles/sec = 2×10^6 cycles/sec). Although to be a successful sonographer may not require mastering problems of this type, nevertheless, we will continue to explore examples such as this throughout the first few chapters of this book in an effort to improve our understanding of the physical factors involved in sound transmission through soft tissue.

The wavelength concept is important in ultrasound physics because it is related to imaging factors such as *spatial resolution*. In addition, the physical size of an object (e.g., a reflecting surface or a transducer surface) is significant only when we compare it to the ultrasonic wavelength. It might be said then that the wavelength is our "acoustic yard-

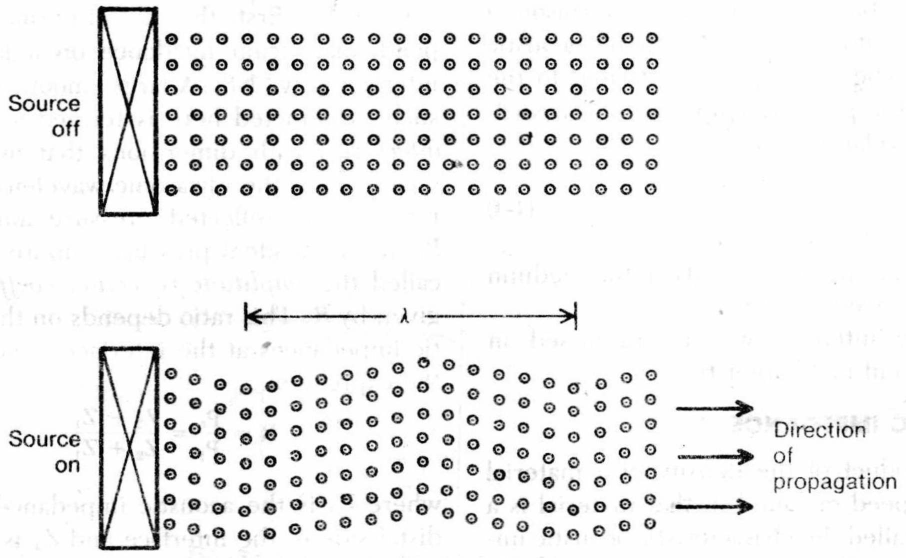

Fig. 1-5. Characteristics of transverse waves, for which the particles in the medium vibrate perpendicular to the direction of propagation of the wave.

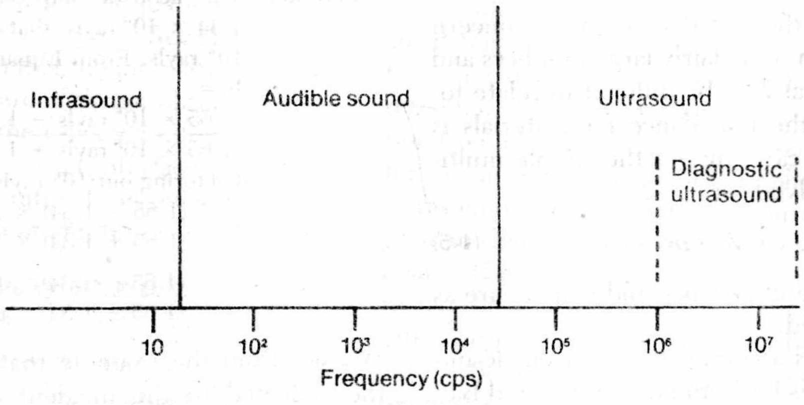

Fig. 1-6. Division of sound into different frequency ranges. *Ultrasound* refers to a sound wave whose frequency is greater than 20 kHz.

stick" (Fig. 1-7). Objects are large or small depending on their size relative to it. In soft tissue, wavelengths for diagnostic ultrasound are on the order of 1 mm or less, with 0.77-mm wavelengths for 2-MHz beams and proportionally smaller ones for higher frequencies.

AMPLITUDE AND INTENSITY

When discussing reflection, attenuation, and scatter, we often must make a quantitative statement regarding the magnitude of a sound wave. One variable that can be used here is the *pressure amplitude*. The acoustic pressure amplitude was illustrated in Fig. 1-1 and was defined as the maximum increase (or decrease) in the pressure relative to ambient conditions in the absence of the sound wave. Other parameters that could have been used in an analogous fashion include the maximum *particle displacement* in the wave and the maximum *particle velocity*.

In some applications, particularly when discussing biologic effects of ultrasound (Chapter 6), it is useful to specify the acoustic intensity. The intensity, I, is related to the square of the pressure amplitude, P, according to the relationship:

$$I = \frac{P^2}{2\rho c} \qquad (1\text{-}4)$$

where, again, ρ is the density of the medium and c the speed of sound.

Acoustic intensity will be discussed in greater detail in Chapter 6.

ACOUSTIC IMPEDANCE

The product of the density of a material and the speed of sound in that material is a quantity called the characteristic acoustic impedance or, for our purposes, simply the *acoustic impedance* of a medium. The significance of this quantity is its role in determining the amplitude of reflected and transmitted waves at an interface. This is discussed in the next section.

Except for the fact that we must concern ourselves with some fairly large numbers and some units that may be difficult to relate to, determining the impedance for materials is just a case of carrying out the simple multiplication involved, or

$$Z = \rho c \qquad (1\text{-}5)$$

where Z is the impedance and ρ and c are as already defined.

Following is a compilation of acoustic impedance values for both nonbiologic and biologic tissues. The units for expressing these are kg/m²/sec (kilograms per square meter per second), which result after multiplying density times speed. Sometimes we find impedance given in *rayls*. One rayl is the same as 1 kg/m²/sec:

	rayls
Air	0.0004×10^6
Lung	0.18×10^6
Fat	1.34×10^6
Water	1.48×10^6
Liver	1.65×10^6
Blood	1.65×10^6
Kidney	1.63×10^6
Muscle	1.71×10^6
Skull bone	7.8×10^6

REFLECTION

Whenever an ultrasound beam is incident on an interface formed by two materials having different acoustic impedances, in general, some of the energy in the beam will be reflected and the remainder transmitted. The amplitude of the reflected wave depends on the difference between the acoustic impedances of the two materials forming the interface.

Consider, first, the case of normal or perpendicular beam incidence on a large flat interface (Fig. 1-8). A large smooth interface such as depicted here is termed a *specular interface*—with dimensions that are much greater than the ultrasonic wavelength. The ratio of the reflected pressure amplitude, P_r, to the incident pressure amplitude, P_i, is called the *amplitude reflection coefficient*—given by R. This ratio depends on the acoustic impedances at the interface according to the expression:

$$R = \frac{P_r}{P_i} = \frac{Z_2 - Z_1}{Z_2 + Z_1} \qquad (1\text{-}6)$$

where Z_2 is the acoustic impedance on the distal side of the interface and Z_1 is the impedance on the proximal side.

Example: Using the values for acoustic impedance just given, calculate the amplitude reflection coefficient for a fat-liver interface.

Solution: The acoustic impedance of fat is 1.34×10^6 rayls, that of liver 1.65×10^6 rayls. From Equation 1-6

$$R = \frac{1.65 \times 10^6 \text{ rayls} - 1.34 \times 10^6 \text{ rayls}}{1.65 \times 10^6 \text{ rayls} + 1.34 \times 10^6 \text{ rayls}}$$

Factoring out 10^6 rayls gives

$$R = \frac{(1.65 - 1.34) \times 10^6 \text{ rayls}}{(1.65 + 1.34) \times 10^6 \text{ rayls}}$$

$$= \frac{(1.65 - 1.34)}{(1.65 + 1.34)} = \frac{0.31}{2.99} = 0.10$$

We see from the example that the ratio of the reflected to the incident amplitude is quite small. In fact, at most soft tissue–soft tissue interfaces in the body the reflection coefficient is fairly small and most of the sound is *transmitted* through the interface. If this were not the case, it would be difficult to use diagnostic ultrasound for examining anatomic structures at significant tissue depths.

Example: Calculate the reflection coefficient for a muscle-air interface.

Solution: From the acoustic impedances given in the list, calculate

$$R = \frac{0.0004 \times 10^6 - 1.7 \times 10^6}{0.0004 \times 10^6 + 1.7 \times 10^6}$$

$$= \frac{0.0004 - 1.7}{0.0004 + 1.7}$$

$$= -0.99$$

(Notice that several of the mathematical steps illustrated in the previous example were combined into one step.) In this case the beam is almost completely reflected. This example illustrates the difficulty in transmitting ultrasound beyond any tissue-to-air interface. Nearly total reflection results in virtually no sound beyond the interface (Fig. 1-9). The complete reflection at air interfaces also explains the need for a coupling medium, such as gel or oil, between the ultrasound transducer (discussed in Chapter 2) and the patient during ultrasound examinations. The coupling material ensures that no air is trapped between the transducer and the skin surface, thereby providing good sound transmission into the patient.

Other examples of reflection coefficients (P_r/P_i) calculated for specular reflecting interfaces are as follows:

Muscle-air	−0.99
Fat-liver	0.10
Kidney-liver	0.006
Liver-muscle	0.018
Muscle-bone	0.64

The data presented here show that a soft tissue–to–bone interface also is a fairly strong reflector. In the majority of ultrasound examinations discussed in this text, bone is avoided because of this and other difficulties associated with propagation through it. Most soft tissue interfaces of importance are fairly weakly reflecting, just as we calculated in the first example.

In summary, reflection of a sound beam occurs whenever the beam is incident on an interface formed by two tissues having different acoustic impedances. The acoustic impedance difference could be caused by a change in speeds of sound, a change in densities, or both. The magnitude of the reflected wave, expressed here as the ratio of the reflected wave amplitude to the incident amplitude, is mainly dependent on the acoustic impedance *difference* at the interface. Interfaces characterized by a large difference in acoustic impedances reflect more of the inci-

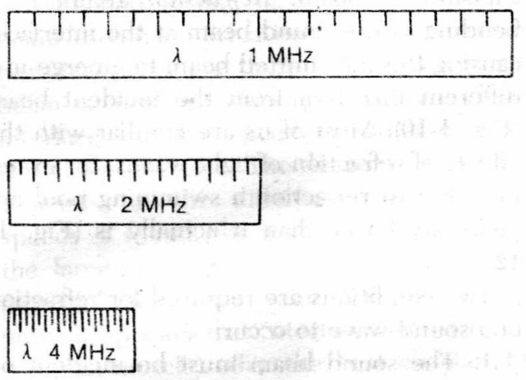

Fig. 1-7. The wavelength is often used as an acoustic yardstick.

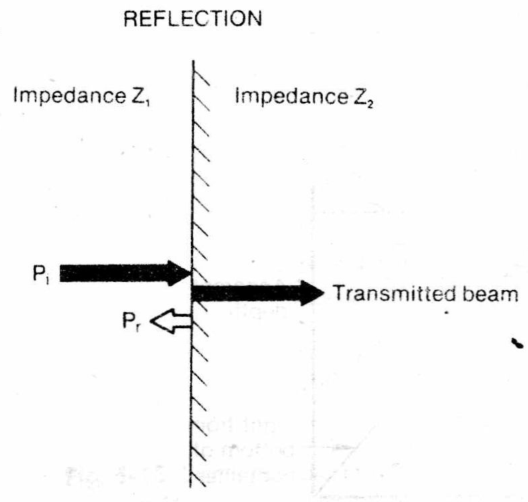

Fig. 1-8. Reflection for perpendicular beam incidence on a specular reflector. P_i is the pressure amplitude of the incident beam, and P_r the amplitude for the reflected beam.

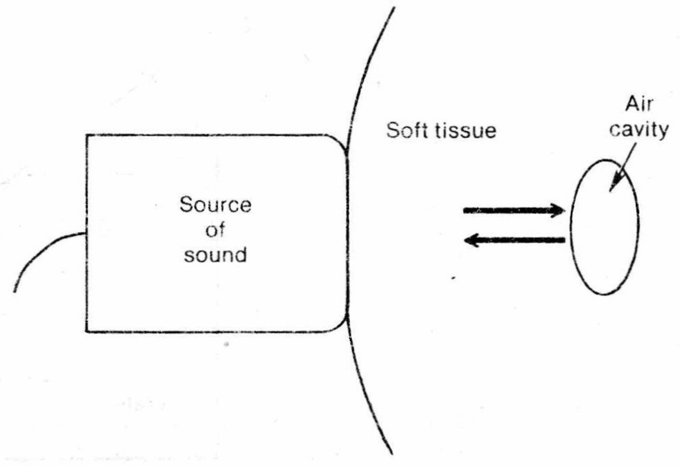

Fig. 1-9. Reflection at a tissue-air interface. Essentially all the sound energy is reflected.

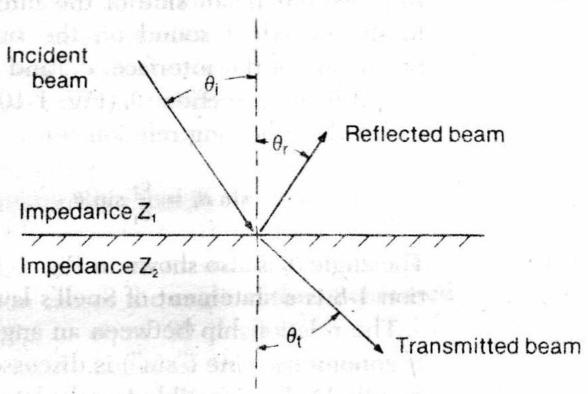

Fig. 1-10. Reflection and refraction for nonperpendicular beam incidence. The incident beam angle, θ_i, reflected beam angle, θ_r, and transmitted beam angle, θ_t, are illustrated.

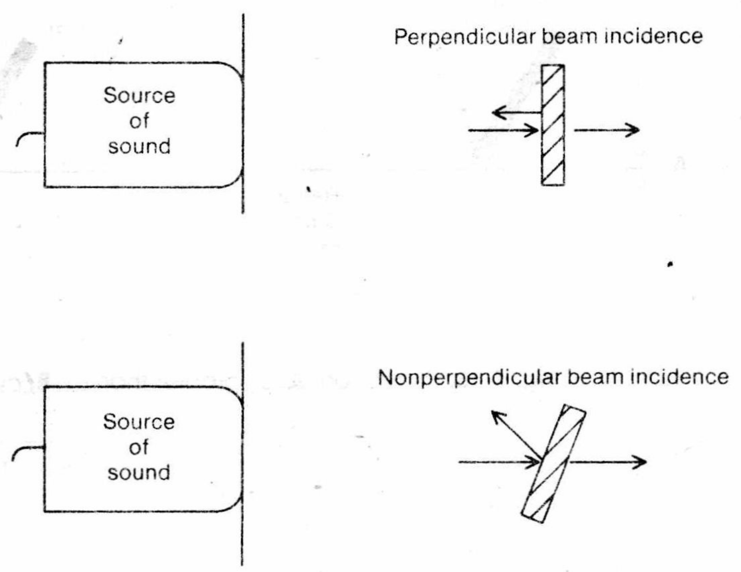

Fig. 1-11. For perpendicular beam incidence the echo returns from a specular reflector toward the source. For nonperpendicular incidence the echo travels in a direction that may miss the source.

dent beam energy than do interfaces where the acoustic impedance difference (*mismatch*) is small.

One additional note: some authors utilize the intensity reflection coefficient rather than the amplitude reflection coefficient to quantify the reflection process. The expression for the size of the reflection looks similar to Equation 1-6, except that the quantity involving the acoustic impedances is squared. In other words, if I_r is the reflected intensity and I_i is the incident intensity, then

$$\frac{I_r}{I_i} = \left[\frac{Z_2 - Z_1}{Z_2 + Z_1}\right]^2 \qquad (1\text{-}7)$$

The two expressions (Equations 1-7 and 1-6) are not contradictory. Recall from our earlier discussion that the intensity is *proportional* to the square of the amplitude. Therefore the ratio of the reflected intensity to the incident intensity at an interface is *equal* to the square of the ratio of the reflected amplitude to the incident amplitude.

NONPERPENDICULAR SOUND BEAM INCIDENCE

For nonperpendicular beam incidence on a specular reflector the situation changes somewhat.

First, the reflected beam does not travel back toward the source (Fig. 1-10) but instead travels off at an angle, θ_r, that is equal to the incident angle, θ_i, only in the opposite direction. This has an effect on echo detection from interfaces. As we shall see in Chapter 3, in many diagnostic applications of ultrasound the sound beam source is also used to detect echoes from reflectors in the beam. The amplitude of an echo that is detected depends on the orientation of the interface relative to the incident beam (Fig. 1-11). Because of this significant angular dependence on the detection of an echo, specular reflectors are sometimes difficult to pick up by a single pulse-echo transducer.

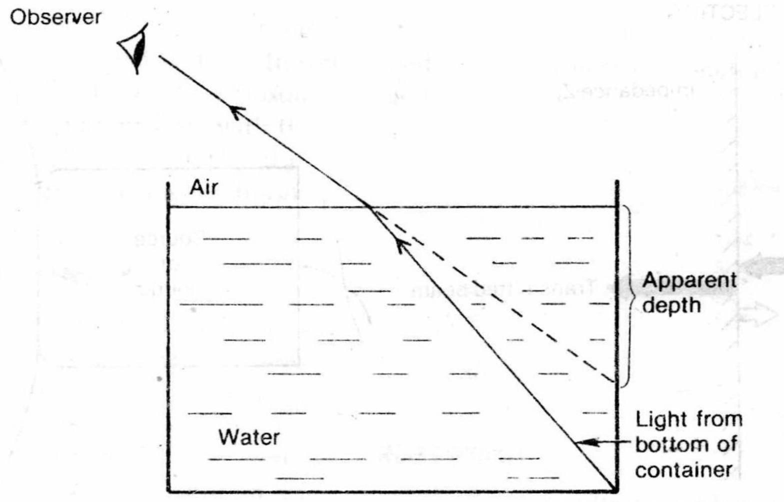

Fig. 1-12. Refraction of light at a water-air interface. To the observer the container of water seems to be shallower than it actually is.

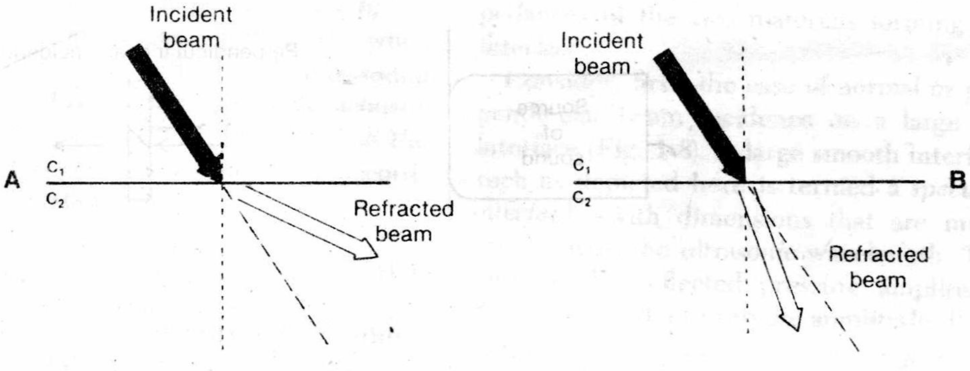

Fig. 1-13. Refraction. **A,** c_2 greater than c_1. **B,** c_2 less than c_1.

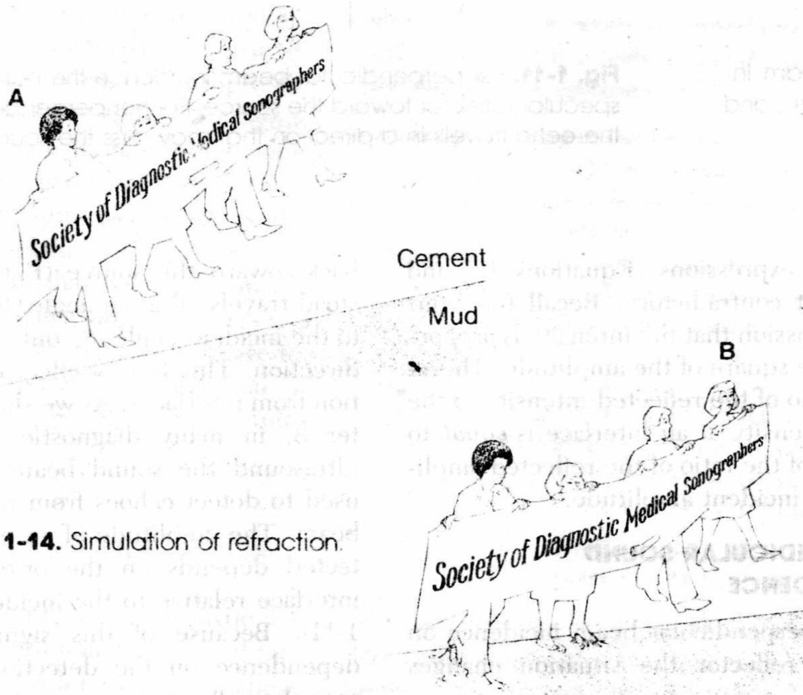

Fig. 1-14. Simulation of refraction.

A second factor that arises when the incident beam is not perpendicular to an interface is the possibility of refraction of the transmitted beam. *Refraction* refers to a bending of the sound beam at the interface, causing the transmitted beam to emerge in a different direction from the incident beam (Fig. 1-10). Most of us are familiar with the effects of refraction of light waves; for example, due to refraction a swimming pool appears shallower than it actually is (Fig. 1-12).

Two conditions are required for refraction of a sound wave to occur:

1. The sound beam must be incident on the interface at an angle that is not perpendicular.
2. The speeds of sound must be different on the two sides of the interface.

Notice what the second condition is saying: it is not sufficient simply to have a reflecting interface to produce refraction; there *also* must be a speed of sound change at the interface for refraction to occur.

The direction of the transmitted (not reflected) beam is governed by Snell's law. The direction is related to the speed of sound on the incident beam side of the interface, c_1, to the speed of sound on the transmitted beam side of the interface, c_2, and to the incident beam direction, θ_i (Fig. 1-10), according to the following relationship:

$$\sin \theta_t = \frac{c_2}{c_1} \sin \theta_i \quad (1\text{-}8)$$

The angle θ_t is also shown in Fig. 1-10. Equation 1-8 is a statement of Snell's law.

The relationship between an angle and its trigonometric sine ("sin") is discussed in Appendix D. It is possible to calculate θ_t, given the incident beam direction and the speeds of sound at the interface. We will not do calculations here using Equation 1-8; suffice to say that the sine of any angle between 0 and 90 degrees increases as the angle itself increases. Therefore, if c_2 is greater than c_1, the angle θ_t will be greater than θ_i; and vice versa (Fig. 1-13). Notice, if c_2 equals c_1, θ_t equals θ_i (i.e., there is no refraction).

To help understand the process of refraction, consider the situation of a row of sonographers carrying a long banner (Fig. 1-14, A). Suppose the sonographers are all walking at the same speed on a concrete pavement as shown. At the end of the pavement is a field of mud, which significantly slows the pace each sonographer can run upon entering it. Then, at some later time, the different speeds that can be maintained on either side of the concrete-mud interface result in the