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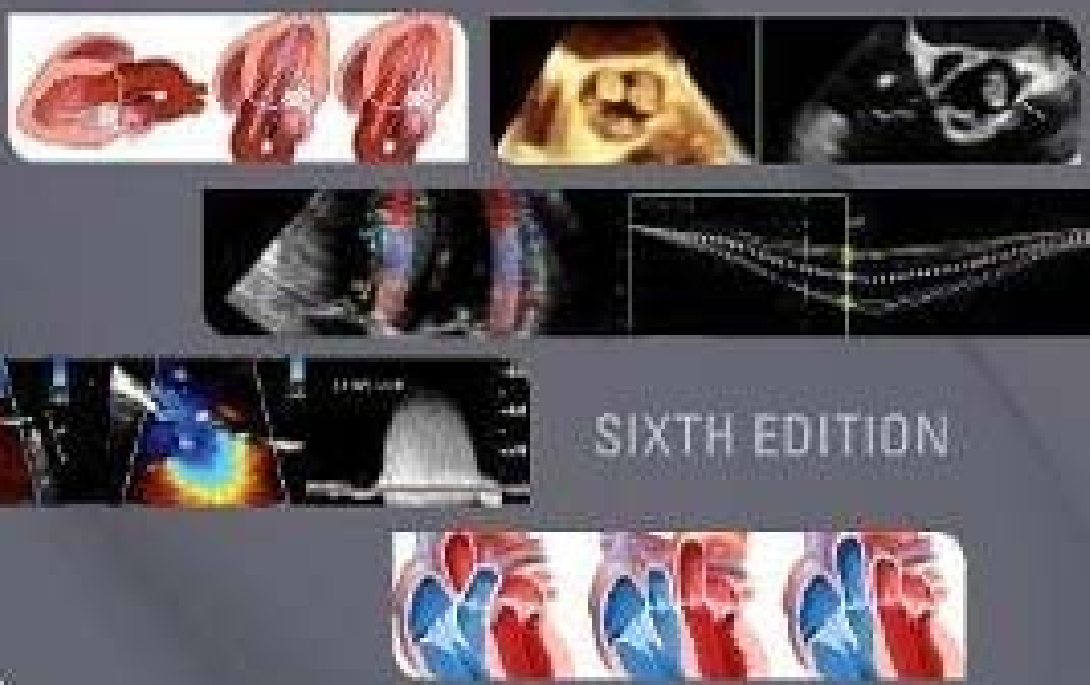
Catherine M. Otto

TEXTBOOK of CLINICAL ECHOCARDIOGRAPHY



SIXTH EDITION

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Catherine M. Otto, MD
J. Ward Kennedy-Hamilton Endowed Chair in Cardiology
Professor of Medicine
University of Washington School of Medicine;
Director, Heart Valve Disease Clinic;
Associate Director, Echocardiography Laboratory
University of Washington Medical Center
Seattle, Washington

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1600 John F. Kennedy Blvd.
Ste 1800
Philadelphia, PA 19103-2899

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PREFACE

Echocardiography is an integral part of clinical cardiology, with important applications in diagnosis, clinical management, and decision making for patients with a wide range of cardiovascular diseases. In addition to examinations performed in the echocardiography laboratory, ultrasound imaging is now used in many other clinical settings, including the emergency department, coronary care unit, intensive care unit, operating room, catheterization laboratory, and electrophysiology laboratory, both for diagnosis and for monitoring the effects of therapeutic interventions. Echocardiographic applications continue to expand because of the detailed and precise anatomic and physiologic information that can be obtained at the bedside with this technique at a relatively low cost and with minimal risk to the patient.

This textbook on general clinical echocardiography is intended to be read by individuals new to echocardiography and by those interested in updating their knowledge in this area. Thus, this book is aimed primarily at cardiology fellows on their basic echocardiography rotation but also will be of value to residents and fellows in general internal medicine, radiology, anesthesiology, and emergency medicine, as well as to cardiac sonography students. For physicians in practice, this textbook provides a concise and practical update.

Textbook of Clinical Echocardiography is structured around a clinical approach to echocardiographic diagnosis. First, a framework of basic principles is provided with chapters on ultrasound physics, normal tomographic transthoracic and transesophageal views, intracardiac flow patterns, indications for echocardiography, and evaluation of left ventricular systolic and diastolic function. A chapter on advanced echocardiographic modalities summarizes basic concepts for 3D echocardiography, myocardial mechanics, contrast echocardiography, and intracardiac echocardiography. Clinical use of these modalities is fully integrated into subsequent chapters, organized by disease categories aligned with the current practice of clinical cardiology.

Each of these chapters summarizes basic principles, the echocardiographic approach, differential diagnosis, technical considerations, and alternate diagnostic approaches. Schematic diagrams illustrate core concepts; echocardiographic images and Doppler recordings show typical findings for each disease process. Transthoracic and transesophageal images, Doppler data, 3D imaging, and other advanced imaging modalities are used throughout the text, reflecting their use in clinical practice. Tables are used frequently to summarize studies validating quantitative echocardiographic methods and to highlight the clinical correlates for each echocardiographic finding. A selected list of annotated references is included at the end of each chapter for those interested in reading more about a particular subject.

Some special features of this book that grew out of my experience teaching physicians and sonographers include The Echo Exam and Echo Math boxes. The Echo Exam consists of concise tables that summarize key concepts at the end of each chapter. Echo Math boxes provide examples of the quantitative calculations used in the day-to-day clinical practice of echocardiography. My hope is that The Echo Exam and Echo Math boxes will serve as quick reference guides in daily practice.

In this sixth edition, each chapter has been revised to reflect advances in the field, suggested readings have been updated, and the majority of the figures have been replaced with recent examples that more clearly illustrate the disease process. Most figures now have a linked video; on your smart device, simply click the video icon to see the echo images in motion. Detailed tables for normal reference values and

evidence tables summarizing validation of quantitative echocardiographic methods are provided in Appendices at the end of the book.

A more advanced discussion of the impact of echocardiographic data in clinical medicine is available in a larger reference book, *The Practice of Clinical Echocardiography*, fifth edition (CM Otto [ed], 2017), also published by Elsevier. Additional clinical examples, practical tips for data acquisition, and multiple-choice self-assessment questions with detailed answers can be found in *Echocardiography Review Guide*, fourth edition, by Freeman, Schwaegler, Linefsky, and Otto (Elsevier, 2018), which exactly parallels the chapters in this textbook.

It should be emphasized that this textbook (or any book) is only a starting point or frame of reference for learning echocardiography. Appropriate training includes competency in the acquisition and interpretation of echocardiographic and Doppler data in real time. Additional training is needed for performance of stress and transesophageal examinations. As echocardiography continues to evolve and new techniques become practical and widely available, practitioners need to update their knowledge. Obviously, a textbook cannot replace the experience gained from performing studies on patients with a range of disease processes, and selected figures with videos do not replace the need for acquisition and review of complete patient examinations. Guidelines for training in echocardiography have been published, as referenced in Chapter 5, and include recommendations for determining clinical competency. Although this textbook is not a substitute for appropriate training and experience, my hope is that it will enhance the learning experience of those new to the field and provide a review for those currently engaged in the acquisition and interpretation of echocardiography. Every patient deserves a clinically appropriate and diagnostically accurate echocardiographic examination; each of us needs to continuously strive toward that goal.

Catherine M. Otto, MD

ACKNOWLEDGMENTS

Many people have provided input to each edition of the *Textbook of Clinical Echocardiography,* and the book is immeasurably enhanced by their contributions—not all can be individually thanked here. First, my special thanks go to the cardiac sonographers at the University of Washington for the outstanding quality of their echocardiographic examinations and for our frequent discussions of the details of image acquisition and the optimal echocardiography examination. Their skill in obtaining superb images provides the basis of many of the figures in this book. My thanks to Pamela Clark, RDCS; Sarah Curtis, RDCS; Maurizio Corona, RDCS; Caryn D'Jang, RDCS; Margaret Falkenreck, RDCS; Michelle Fujioka, RDCS; Carolyn Gardner, RDCS; Deanna Knox, RDCS; Yelena Kovolenko, RDCS; Carol Kraft, RDCS; Carin Lodell, RDCS; Chris McKenzie, RDCS; Irina Nesterova, RDCS; Amy Owens, RDCS; Becky Schwaegler, RDCS; Joanna Shephard, RDCS; Karl Skinner, RDCS; Hoang Tran, RDCS; Yu Wang, RDCS; and Todd Zwink, RDCS.

My gratitude extends to my colleagues at the University of Washington who shared their expertise and helped identify images for the book, with special thanks to Rosario Freeman, MD; Jim Kirkpatrick, MD; Don Oxorn, MD; and G. Burkhard Mackensen, MD. The University of Washington Cardiology Fellows also provided thoughtful (and sometimes humbling) insights and helped identify cases with teaching quality images; particular recognition goes to James C. Lee. In addition, my gratitude is extended to those individuals who kindly gave permission for reproduction of previously published figures. Joe Chovan and Starr Kaplan are to be commended for their skills as medical illustrators and for providing such clear and detailed anatomic drawings.

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Many thanks to my editor, Delores Meloni, at Elsevier for providing the support needed to write this edition and to Jennifer Ehlers and the production team for all the detail-oriented hard work that went into making this book a reality.

Finally, my most appreciative thanks to my husband, daughter, and granddaughter for their unwavering support in every aspect of life.

Catherine M. Otto, MD

KEY EQUATIONS

Ultrasound Physics	
Frequency	f = cycles/s = Hz
Wavelength	$\lambda = c / f = 1.54 / f$ (MHz)
Doppler equation	$v = c \times \Delta f / [2F_T (\cos\theta)]$
Bernoulli equation	$\Delta P = 4V^2$
LV Imaging	
Stroke volume	
Ejection fraction	EF(%) = (SV / EDV) × 100%
Wall stress	σ = PR / 2Th
Doppler Ventricular Function	
Stroke volume	SV = CSA × VTI
Rate of pressure rise	dP/dt = 32 mmHg / Time from 1 to 3 m/s of MR CW jet(sec)
Myocardial performance index	MPI = (IVRT + IVCT) / SEP
Pulmonary Pressures and Resistance	
Pulmonary systolic pressure	$PAP_{systolic} = 4(V_{TR})2 + RAP$
PAP (when PS is present)	$PAP_{systolic} = [4(V_{TR})2 + RAP] - \Delta P_{RV–PA}$
Mean PA pressure	PAP_{mean} = Mean $\Delta P_{RV} -$ $_{RA}$ + RAP
Diastolic PA pressure	$PAP_{diastolic} = 4 (V_{PR})^2 + RAP$
Pulmonary vascular resistance	$PVR \approx 10(V_{TR}) / VTI_{RVOT}$
Aortic Stenosis	
Maximum pressure gradient (integrate over ejection period for mean gradient)	$\Delta P_{max} = 4(V_{max})^2$
Continuity equation valve area	$AVA(cm^2) = [\pi(LVOT_D / 2)^2 \times VTI_{LVOT}] / VTI_{AS\text{-}Jet}$
Simplified continuity equation	$AVA(cm^2) = [\pi (LVOT_D / 2)^2 \times V_{LVOT}] / V_{AS\text{-}Jet}$
Velocity ratio	Velocity ratio = $V_{LVOT} / V_{AS\text{-}Jet}$
Mitral Stenosis	
Pressure half-time valve area	$MVA_{Doppler} = 220 / T½$
Aortic Regurgitation	
Total stroke volume	$TSV = SV_{LVOT} = (CSA_{LVOT} \times VTI_{LVOT})$
Forward stroke volume	$FSV = SV_{MA} = (CSA_{MA} \times VTI_{MA})$
Regurgitant volume	RVol = TSV – FSV
Regurgitant orifice area	$ROA = RSV / VTI_{AR}$
Mitral Regurgitation	
Total stroke volume (or 2D or 3D LV stroke volume)	$TSV = SV_{MA} = (CSA_{MA} \times VTI_{MA})$
Forward stroke volume	$FSV = SV_{LVOT} = (CSA_{LVOT} \times VTI_{LVOT})$
Regurgitant volume	RVol = TSV – FSV
Regurgitant orifice area	$ROA = RSV / VTI_{AR}$
PISA Method	
Regurgitant flow rate	$R_{FR} = 2\pi r^2 \times V_{aliasing}$
Orifice area (maximum)	$ROA_{max} = R_{FR} / V_{MR}$
Regurgitant volume	$RV = ROA \times VTI_{MR}$
Aortic Dilation	
Predicted sinus diameter	
Children (<18 years): Predicted sinus dimension = 1.02 + (0.98 BSA)	
Adults (18–40 years): Predicted sinus dimension = 0.97 + (1.12 BSA)	
Adults (>40 years): Predicted sinus dimension = 1.92 + (0.74 BSA)	
Ratio = Measured maximum diameter / Predicted maximum diameter	
Pulmonary (Q_p) to Systemic (Q_s) Shunt Ratio	
$Q_p:Q_s$	$= [CSA_{PA} \times VTI_{PA}] / [CSA_{LVOT} \times VTI_{LVOT}]$

1 Principles of Echocardiographic Image Acquisition and Doppler Analysis

An understanding of the basic principles of ultrasound imaging and Doppler echocardiography is essential both during data acquisition and for correct interpretation of the ultrasound information. Although, at times, current instruments provide instantaneous images so clear and detailed that it seems as if we can "see" the heart and blood flow directly, in actuality we always are looking at images and flow data generated by complex analyses of ultrasound waves reflected and backscattered from the patient's body. Knowledge of the strengths, and more importantly the limitations, of this technique is critical for correct clinical diagnosis and patient management. On the one hand, echocardiography can be used for decision making with a high degree of accuracy in a variety of clinical settings. On the other hand, if an ultrasound artifact is mistaken for an anatomic abnormality, a patient could undergo other needless, expensive, and potentially risky diagnostic tests or therapeutic interventions.

In this chapter, a brief (and necessarily simplified) overview of the basic principles of cardiac ultrasound imaging and flow analysis is presented. The reader is referred to the Suggested Reading at the end of the chapter for more information on these subjects. Because the details of image processing, artifact formation, and Doppler physics become more meaningful with experience, some readers may choose to return to this chapter after reading other sections of this book and after participating in some echocardiographic examinations.

ULTRASOUND WAVES

Sound waves are mechanical vibrations that induce alternate refraction and compression of any physical medium through which they pass (Fig. 1.1). Like other waves, sound waves are described in terms of (Table 1.1):

- Frequency: cycles per second, or hertz (Hz)
- Velocity of propagation
- Wavelength: millimeters (mm)
- Amplitude: decibels (dB)

Frequency (f) is the number of ultrasound waves in a 1-second interval. The units of measurement are hertz, abbreviated Hz, which simply means cycles per second. A frequency of 1000 cycles/s is 1 kilohertz (kHz), and 1 million cycles/s is 1 megahertz (MHz). Humans can hear sound waves with frequencies between 20 Hz and 20 kHz; frequencies higher than this range are termed *ultrasound*. Diagnostic medical ultrasound typically uses transducers with a frequency between 1 and 20 MHz.

The speed that a sound wave moves through the body, called the *velocity of propagation* (*c*), is different for each type of tissue. For example, the velocity of propagation in bone is much faster (about 3000 m/s) than in lung tissue (about 700 m/s). However, the velocity of propagation in soft tissues, including myocardium, valves, blood vessels, and blood, is relatively uniform, averaging about 1540 m/s.

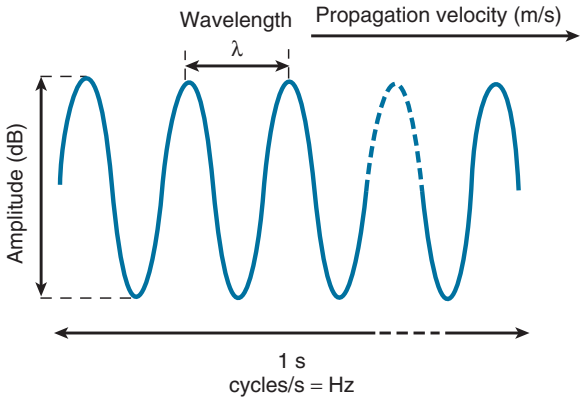

Fig. 1.1 Schematic diagram of an ultrasound wave.

ECHO MATH: Wavelength

Wavelength (**λ**) is the distance from peak to peak of an ultrasound wave. Wavelength can be calculated by dividing the frequency (*f* in Hz) by the propagation velocity (*c* in m/s):

$$\lambda = c/f \qquad \text{(Eq. 1.1)}$$

Because the propagation velocity in the heart is constant at 1540 m/s, the wavelength for any transducer frequency can be calculated (Fig. 1.2) as:

$$\lambda\,(\text{mm}) = [1540\ \text{m/s}/f\,(\text{Hz})]/1000\ \text{mm/m}$$

or

$$\lambda\,(\text{mm}) = 1.54/f$$

For example, the wavelength emitted by a 5-MHz transducer can be calculated as:

$$\lambda = 1540\ \text{m/s}/5{,}000{,}000\ \text{cycle/s} = 0.000308\ \text{m} = 0.308\ \text{mm}$$

TABLE 1.1 Ultrasound Waves

	Definition	Examples	Clinical Implications
Frequency (*f*)	The number of cycles per second in an ultrasound wave *f* = cycles/s = Hz	Transducer frequencies are measured in MHz (1,000,000 cycles/s). Doppler signal frequencies are measured in kHz (1000 cycles/s).	Different transducer frequencies are used for specific clinical applications because the transmitted frequency affects ultrasound tissue penetration, image resolution, and the Doppler signal.
Velocity of propagation (c)	The speed that ultrasound travels through tissue	The average velocity of ultrasound in soft tissue is ≈1540 m/s.	The velocity of propagation is similar in different soft tissues (e.g., blood, myocardium, liver, fat) but is much lower in lung and much higher in bone.
Wavelength (λ)	The distance between ultrasound waves: λ = c/*f* = 1.54 / *f* (MHz)	Wavelength is shorter with a higher-frequency transducer and longer with a lower-frequency transducer.	Image resolution is greatest (≈1 mm) with a shorter wavelength (higher frequency). Depth of tissue penetration is greatest with a longer wavelength (lower frequency).
Amplitude (dB)	Height of the ultrasound wave or "loudness" measured in decibels (dB)	A log scale is used for dB. On the dB scale, 80 dB represents a 10,000-fold and 40 dB indicates a 100-fold increase in amplitude.	A very wide range of amplitudes can be displayed using a gray-scale display for both imaging and spectral Doppler.

Wavelength is important in diagnostic applications for at least two reasons:

- Image resolution is no greater than 1 to 2 wavelengths (typically about 1 mm).
- The depth of penetration of the ultrasound wave into the body is directly related to wavelength; shorter wavelengths penetrate a shorter distance than longer wavelengths.

Thus there is an obvious trade-off between image resolution (shorter wavelength or higher frequency preferable) and depth penetration (longer wavelength or lower frequency preferable).

The acoustic pressure, or amplitude, of an ultrasound wave indicates the energy of the ultrasound signal. Power is the amount of energy per unit time. Intensity (I) is the amount of power per unit area:

$$\text{Intensity (I)} = \text{Power}^2 \qquad \text{(Eq. 1.2)}$$

This relationship shows that if ultrasound power is doubled, intensity is quadruped. Instead of using direct measures of pressure energy, ultrasound amplitude is described relative to a reference value using the decibel scale. Decibels (dB) are familiar to all of us as the standard description of the loudness of a sound.

ECHO MATH: Decibels

Decibels are logarithmic units based on a ratio of the measured amplitude (A_2) to a reference amplitude (A_1) such that:

$$dB = 20 \log(A_2/A_1) \qquad \text{(Eq. 1.3)}$$

Thus a ratio of 1000 to 1 is

$$20 \times \log(1000) = 20 \times 3 = 60 \text{ dB}$$

a ratio of 100 to 1 is

$$20 \times \log(100) = 20 \times 2 = 40 \text{ dB}$$

and a ratio of 2 to 1 is

$$20 \times \log(2) = 20 \times 0.3 = 6 \text{ dB}$$

A simple rule to remember is that a 6-dB change represents a doubling or halving of the signal amplitude or that a 40-dB change represents a 100 times difference in amplitude (Fig. 1.3).

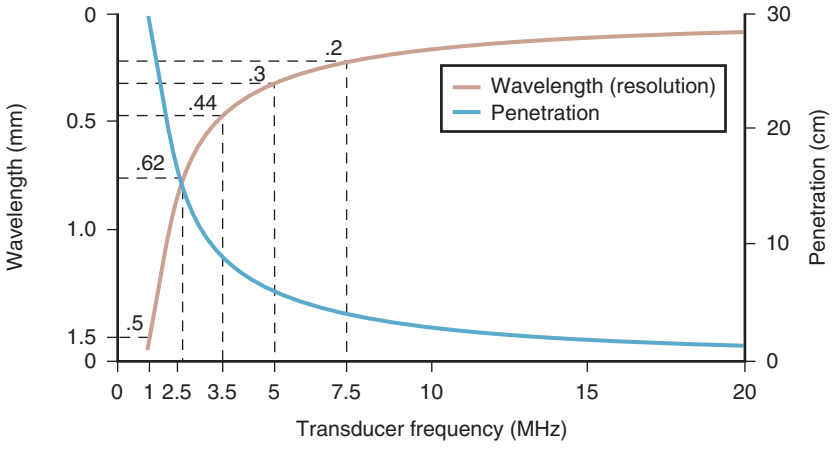

Fig. 1.2 Transducer frequency versus wavelength and penetration of the ultrasound signal in soft tissue. Wavelength has been plotted inversely to show that resolution increases with increasing transducer frequency while penetration decreases. The specific wavelengths for transducer frequencies of 1, 2.5, 3.5, 5, and 7.5 MHz are shown.

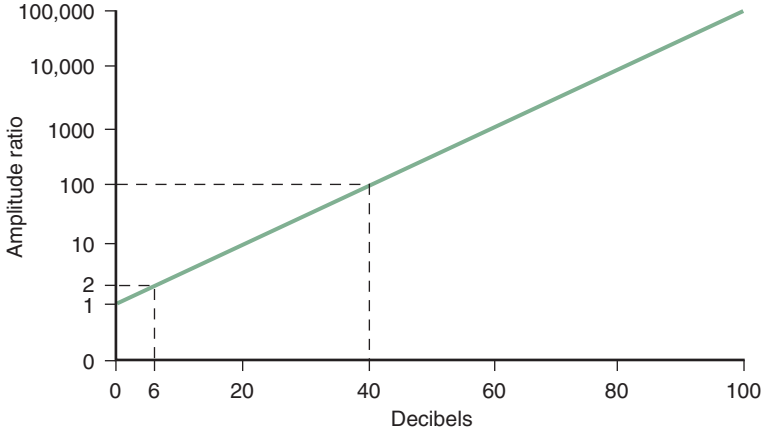

Fig. 1.3 Graph of the decibel scale. The logarithmic relationship between the decibel scale (horizontal axis) and the amplitude ratio (vertical axis) is seen. A doubling or halving of the amplitude ratio corresponds to a 6-dB change, and a 100-fold difference in amplitude corresponds to a 20-dB change.

If acoustic intensity is used instead of amplitude, the constant 10 replaces 20 in the equation so that a 3-dB change represents doubling, and a 20-dB change indicates a 100-fold difference in amplitude. Both these decibel scales are used to refer to transmitted or received ultrasound waves or to describe attenuation effects. The advantages of the decibel scale are that a very large range can be compressed into a smaller number of values and that low-amplitude (weak) signals can be displayed alongside very high-amplitude (strong) signals. In an echocardiographic image, amplitudes typically range from 1 to 120 dB. The decibel scale is the standard format both for echocardiographic image display and for the Doppler spectral display, although other amplitude scales are sometimes available.

ULTRASOUND TISSUE INTERACTION

Propagation of ultrasound waves in the body to generate ultrasound images and Doppler data depends on a tissue property called *acoustic impedance* (Table 1.2). Acoustic impedance (Z) depends on tissue density (ρ) and on the propagation velocity in that tissue (c):

$$Z = \rho c \quad \text{(Eq. 1.4)}$$

Although the velocity of propagation differs among tissues, tissue density is the primary determinant of acoustic impedance for diagnostic ultrasound. Lung tissue has a very low density compared with bone, which has a very high density. Soft tissues, such as blood and myocardium, have much smaller differences in tissue density and acoustic impedance. Acoustic impedance determines the transmission of ultrasound waves through a tissue; *differences* in acoustic impedance result in reflection of ultrasound waves at tissue boundaries.

The interaction of ultrasound waves with the organs and tissues of the body can be described in terms of (Fig. 1.4):

- Reflection
- Scattering
- Refraction
- Attenuation

TABLE 1.2 Ultrasound Tissue Interaction

	Definition	Examples	Clinical Implications
Acoustic impedance (Z)	A characteristic of each tissue defined by tissue density (ρ) and propagation of velocity (c) as: $Z = \rho \times c$	Lung has a low density and slow propagation velocity, whereas bone has a high density and fast propagation velocity. Soft tissues have smaller differences in tissue density and acoustic impedance.	Ultrasound is reflected from boundaries between tissues with differences in acoustic impedance (e.g., blood vs. myocardium).
Reflection	Return of ultrasound signal to the transducer from a smooth tissue boundary	Reflection is used to generate 2D cardiac images.	Reflection is greatest with the ultrasound beam in perpendicular to the tissue interface.
Scattering	Radiation of ultrasound in multiple directions from a small structure (e.g., blood cells)	The change in frequency of signals scattered from moving blood cells is the basis of Doppler ultrasound.	The amplitude of scattered signals is 100 to 1000 times less than reflected signals.
Refraction	Deflection of ultrasound waves from a straight path due to differences in acoustic impedance	Refraction is used in transducer design to focus the ultrasound beam.	Refraction in tissues results in double image artifacts.
Attenuation	Loss in signal strength due to absorption of ultrasound energy by tissues	Attenuation is frequency dependent with greater attenuation (less penetration) at higher frequencies.	A lower-frequency transducer is needed for apical views or in larger patients on transthoracic imaging.
Resolution	The smallest resolvable distance between two specular reflectors on an ultrasound image	Resolution has three dimensions: along the length of the beam (axial), lateral across the image (azimuthal), and in the elevational plane.	Axial resolution is most precise (as small as 1 mm), so imaging measurements are best made along the length of the ultrasound beam.

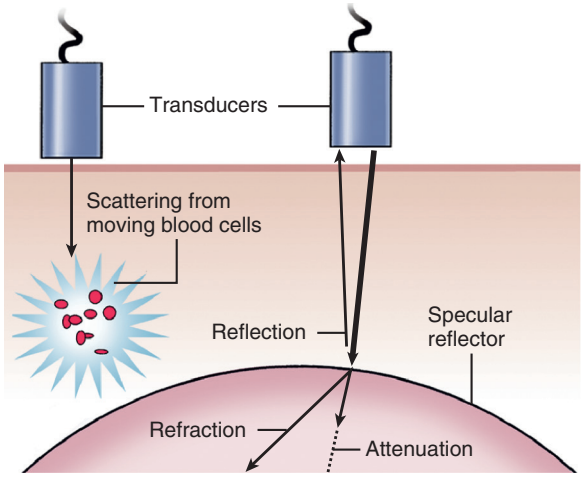

Fig. 1.4 Diagram of the interaction between ultrasound and body tissues. Doppler analysis is based on the scattering of ultrasound in all directions from moving blood cells with a resulting change in frequency of the ultrasound received at the transducer. 2D imaging is based on reflection of ultrasound from tissue interfaces (specular reflectors). Attenuation limits the depth of ultrasound penetration. Refraction, a change in direction of the ultrasound wave, results in imaging artifacts.

Reflection

The basis of ultrasound imaging is *reflection* of the transmitted ultrasound signal from internal structures. Ultrasound is reflected at tissue boundaries and interfaces, with the amount of ultrasound reflected dependent on the:

- Difference in acoustic impedance between the two tissues
- Angle of reflection

Smooth tissue boundaries with a lateral dimension greater than the wavelength of the ultrasound beam act as specular, or "mirror-like," reflectors. The amount of ultrasound reflected is constant for a given interface, although the amount received back at the transducer varies with angle because (like light reflected from a mirror) the angle of incidence and reflection is equal. Thus optimal return of reflected ultrasound occurs at a perpendicular angle (90°). Remembering this fact is crucial for obtaining diagnostic ultrasound images. It also accounts for ultrasound "dropout" in a two-dimensional (2D) or three-dimensional (3D) image when too little or no reflected ultrasound reaches the transducer resulting from a parallel alignment between the ultrasound beam and tissue interface.

Scattering

Scattering of the ultrasound signal, instead of reflection, occurs with small structures, such as red blood cells suspended in fluid, because the radius of the cell (about 4 μm) is smaller than the wavelength of the ultrasound signal. Unlike a reflected beam, scattered ultrasound energy radiates in all directions. Only a small amount of the scattered signal reaches the receiving transducer, and the amplitude of a scattered signal is 100 to 1000 times (40–60 dB) less than the amplitude of the returned signal from a specular reflector. Scattering of ultrasound from moving blood cells is the basis of Doppler echocardiography.

The *extent* of scattering depends on:

- Particle size (red blood cells)
- Number of particles (hematocrit)
- Ultrasound transducer frequency
- Compressibility of blood cells and plasma

Although experimental studies show differences in backscattering with changes in hematocrit, variation over the clinical range has little effect on the Doppler signal. Similarly, the size of red blood cells and the compressibility of blood cells and plasma do not change significantly. Thus the primary determinant of scattering is transducer frequency.

Scattering also occurs within tissues, such as the myocardium, from interference of backscattered signals from tissue interfaces smaller than the ultrasound wavelength. Tissue scattering results in a pattern of *speckles;* tissue motion can be measured by tracking these speckles from frame to frame, as discussed in Chapter 4.

Refraction

Ultrasound waves can be *refracted*—deflected from a straight path—as they pass through a medium with a different acoustic impedance. Refraction of an ultrasound beam is analogous to refraction of light waves as they pass through a curved glass lens (e.g., prescription eyeglasses). Refraction allows enhanced image quality by using acoustic "lenses" to focus the ultrasound beam. However, refraction also occurs in unplanned ways during image formation, with resulting ultrasound artifacts, most notably the "double-image" artifact.

Attenuation

Attenuation is the loss of signal strength as ultrasound interacts with tissue. As ultrasound penetrates into the body, signal strength is progressively *attenuated* because of absorption of the ultrasound energy by conversion to heat, as well as by reflection and scattering. The degree of attenuation is related to several factors, including the:

- Attenuation coefficient of the tissue
- Transducer frequency
- Distance from the transducer
- Ultrasound intensity (or power)

The attenuation coefficient (α) for each tissue is related the decrease in ultrasound intensity (measured

in dB) from one point (I_1) to a second point (I_2) separated by a distance *(l)* as described by the equation:

$$I_2 = I_1 e^{-2\alpha l} \quad \text{(Eq. 1.5)}$$

The attenuation coefficient for air is very high (about 1000×) compared with soft tissue so that any air between the transducer and heart results in substantial signal attenuation. This is avoided on transthoracic examinations by use of a water-soluble gel to form an airless contact between the transducer and the skin; on transesophageal echocardiography (TEE) examination, attenuation is avoided by maintaining close contact between the transducer and esophageal wall. The air-filled lungs are avoided by careful patient positioning and the use of acoustic "windows" that allow access of the ultrasound beam to the cardiac structures without intervening lung tissue. Other intrathoracic air (e.g., pneumomediastinum, residual air after cardiac surgery) also results in poor ultrasound tissue penetration because of attenuation, resulting in suboptimal image quality.

The power output of the transducer is directly related to the overall degree of attenuation. However, an increase in power output causes thermal and mechanical bioeffects as discussed in "Bioeffects and Safety," p. 27.

Overall attenuation is frequency dependent such that lower ultrasound frequencies penetrate deeper into the body than higher frequencies. The depth of penetration for adequate imaging tends to be limited to approximately 200 wavelengths. This translates roughly into a penetration depth of 30 cm for a 1-MHz transducer, 6 cm for a 5-MHz transducer, and 1.5 cm for a 20-MHz transducer, although diagnostic images at depths greater than these postulated limits can be obtained with state-of-the-art equipment. Thus attenuation, as much as resolution, dictates the need for a particular transducer frequency in a specific clinical setting. For example, visualization of distal structures from the apical approach in a large adult patient often requires a low-frequency transducer. From a TEE approach, the same structures can be imaged (at better resolution) with a higher-frequency transducer. The effects of attenuation are minimized on displayed images by using different gain settings at each depth, an instrument control called time-gain (or depth-gain) compensation.

TRANSDUCERS

Piezoelectric Crystal

Ultrasound transducers use a piezoelectric crystal both to generate and to receive ultrasound waves (Fig. 1.5). A piezoelectric crystal is a material (e.g., quartz or a titanate ceramic) with the property that an applied electric current results in alignment of polarized

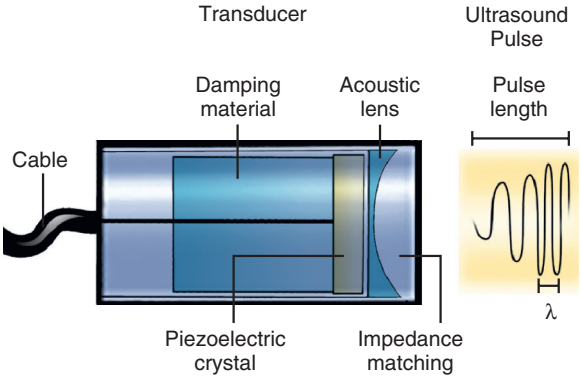

Fig. 1.5 Schematic diagram of an ultrasound transducer. The piezoelectric crystal both produces and receives ultrasound signals, with the electric input-output transmitted to the instrument via the cable. Damping material allows a short pulse length (improved resolution). The shape of the piezoelectric crystal, an acoustic lens, or electronic focusing (with a phased-array transducer) is used to modify beam geometry. The material of the transducer surface provides impedance matching with the skin. The ultrasound pulse length for 2D imaging is short (1–6 ms), typically consisting of two wavelengths (λ). "Ring down"—the decrease in frequency and amplitude in the pulse—depends on damping and determines bandwidth (the range of frequencies in the signal).

particles perpendicular to the face of the crystal with consequent expansion of crystal size. When an alternating electric current is applied, the crystal alternately compresses and expands, generating an ultrasound wave. The frequency that a transducer emits depends on the nature and thickness of the piezoelectric material.

Conversely, when an ultrasound wave strikes the piezoelectric crystal, an electric current is generated. Thus the crystal can serve both as a "receiver" and as a "transmitter." Basically, the ultrasound transducer transmits a brief burst of ultrasound and then switches to the "receive mode" to await the reflected ultrasound signals from the intracardiac acoustic interfaces. This cycle is repeated temporally and spatially to generate ultrasound images. Image formation is based on the *time delay* between ultrasound transmission and return of the reflected signal. Deeper structures have a longer time of flight than shallower structures, with the exact depth calculated based on the speed of sound in blood and the time interval between the transmitted burst of ultrasound and return of the reflected signal.

The burst, or pulse, of ultrasound generated by the piezoelectric crystal is very brief, typically 1 to 6 μs, because a short pulse length results in improved axial (along the length of the beam) resolution. Damping material is used to control the ring-down time of the crystal and hence the pulse length. Pulse length also is determined by frequency because a shorter time is needed for the same number of cycles at higher frequencies. The number of ultrasound pulses per second is called the *pulse repetition frequency*

(PRF). The total time interval from pulse to pulse is called the *cycle length*, with the percentage of the cycle length used for ultrasound transmission called the *duty factor*. Ultrasound imaging has a duty factor of about 1% compared with 5% for pulsed Doppler and 100% for continuous-wave (CW) Doppler. The duty factor is a key element in the patient's total ultrasound exposure.

The range of frequencies contained in the pulse is described as its *frequency bandwidth*. A wider bandwidth allows better axial resolution because of the ability of the system to produce a narrow pulse. Transducer bandwidth also affects the range of frequencies that can be detected by the system with a wider bandwidth, which allows better resolution of structures distant from the transducer. The stated frequency of a transducer represents the center frequency of the pulse.

Types of Transducers

The simplest type of ultrasound transducer is based on a single piezoelectric crystal (Table 1.3). Alternate pulsed transmission and reception periods allow repeated sampling along a single line, with the sampling rate limited only by the time delay needed for return of the reflected ultrasound wave from the depth of interest. An example of using the transducer for simple transmission and reception along a single line is an A-mode (amplitude vs. depth) or M-mode (depth vs. time) cardiac recording when a high sampling rate is desirable.

Formation of more complex images uses an array of ultrasound crystals arranged to provide a 2D tomographic or 3D volumetric data set of signals. Each element in the transducer array can be controlled electronically both to direct the ultrasound beam across the region of interest and to focus the transmitted and received signals. Echocardiographic imaging uses a *sector scanning* format with the ultrasound signal originating from a single location (the narrow end of the sector), thus resulting in a fanlike shape of the image. Sector scanning is optimal for cardiac applications because it allows a fast frame rate to show cardiac motion and a small transducer size (aperture or "footprint") to fit into the narrow acoustic windows used for echocardiography. 3D imaging transducers are discussed in Chapter 4.

Most transducers can provide simultaneous imaging and Doppler analysis, for example, 2D imaging and a superimposed color Doppler display. Quantitative Doppler velocity data are recorded with the image "frozen" or with only intermittent image updates, with the ultrasound crystals used to optimize the Doppler signal. Although CW Doppler signals can be obtained using two elements of combined transducer, use of a dedicated nonimaging transducer with two separate crystals (with one crystal continuously transmitting and the other continuously receiving the ultrasound waves) is recommended when accurate high-velocity recordings are needed. The final configuration of a transducer depends on transducer frequency (higher-frequency transducers are smaller) and beam focusing, as well as the intended clinical use, for example, transthoracic versus TEE imaging.

Beam Shape and Focusing

An unfocused ultrasound beam is shaped like the light from a flashlight, with a tubular beam for a short distance that then diverges into a broad cone of light (Fig. 1.6). Even with current focused transducers, ultrasound beams have a 3D shape that affects measurement accuracy and contributes to imaging artifacts. Beam shape and size depend on several factors, including:

- Transducer frequency
- Distance from the transducer
- Aperture size and shape
- Beam focusing

Aperture size and shape and beam focusing can be manipulated in the design of the transducer, but the effects of frequency and depth are inherent to ultrasound physics. For an unfocused beam, the initial segment of the beam is columnar in shape (near-field F_n) with a length dependent on the diameter (D) of the transducer face and wavelength (λ):

$$F_n = D^2/4\lambda \quad \text{(Eq. 1.6)}$$

For a 3.5-MHz transducer with a 5-mm diameter aperture, this corresponds to a columnar length of 1.4 cm. Beyond this region, the ultrasound beam diverges (far field), with the angle of divergence θ determined as:

$$\sin\theta = 1.22\lambda/D \quad \text{(Eq. 1.7)}$$

This equation indicates a divergence angle of 6° beyond the near field, resulting in an ultrasound beam width of about 4.4 cm at a depth of 20 cm for this 3.5-MHz transducer. With a 10-mm diameter aperture, F_n would be 5.7 cm, and beam width at 20 cm would be about 2.5 cm (Fig. 1.7).

The shape and focal depth (narrowest point) of the primary beam can be altered by making the surface of the piezoelectric crystal concave or by the addition of an acoustic lens. This allows generation of a beam with optimal characteristics at the depth of most cardiac structures, but again, divergence of the beam beyond the focal zone occurs. Some transducers allow manipulation of the focal zone during the examination. Even with focusing, the ultrasound beam generated by each transducer has a lateral and an elevational dimension that depends on the transducer aperture, frequency, and focusing. Beam geometry for phased-array transducers also depends