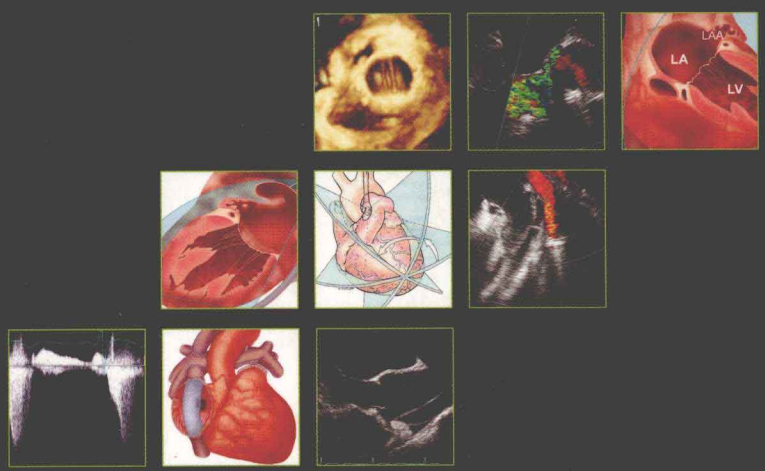
OTTO

*textbook of*

# Clinical Echocardiography

FOURTH EDITION

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FOURTH EDITION

# TEXTBOOK *of* CLINICAL ECHOCARDIOGRAPHY


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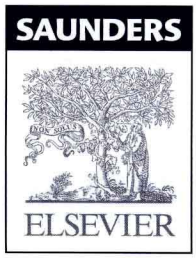

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# PREFACE

Echocardiography is an integral part of clinical cardiology with important applications in the initial diagnosis, clinical management, and decision making for patients with a wide range of cardiovascular diseases. In addition to examinations performed in the echocardiography laboratory, echocardiographic techniques are now used in a variety of other clinical settings, including the coronary care unit, intensive care unit, operating room, emergency department, catheterization laboratory, and electrophysiology laboratory, both for diagnosis and for monitoring the effects of therapeutic interventions. There continues to be expansion of echocardiographic applications given the detailed and precise anatomic and physiologic information that can be obtained 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. The text 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. For additional clinical examples, practical tips for data acquisition, and self-assessment questions, the *Echocardiography Review Guide*, by Schwaegler and Otto (Elsevier/Saunders 2007) parallels the information provided in this textbook. 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, 3rd edition* (CM Otto [ed], 2007), also published by Elsevier/Saunders. The DVD that accompanies that book includes cases with cine images and interactive multiple choice questions for each chapter.

This book 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 introduces the concepts of 3D echocardiography, myocardial mechanics, contrast echocardiography, and intracardiac echocardiography. Clinical use of these modalities is integrated into subsequent chapters as appropriate. Some of these modalities, such as intracardiac and intravascular echocardiography, typically are utilized by cardiologists with training in interventional procedures. Physicians and sonographers who plan to utilize these modalities in their clinical practices are referred to chapters in *The Practice of Clinical Echocardiography* and other suggested reading.

This framework of basic principles then is built upon in subsequent chapters, organized by disease category (for example, cardiomyopathy or valvular stenosis), corresponding to the typical indications for echocardiography in clinical practice. In each chapter, basic principles for echocardiographic evaluation of that disease category are reviewed, the echocardiographic approach and differential diagnosis are discussed in detail, limitations and technical considerations are emphasized, and alternate diagnostic approaches are delineated. Schematic diagrams are used to illustrate basic concepts; echocardiographic images and Doppler data show typical and unusual findings in patients with each disease process. Transthoracic and transesophageal images, Doppler data, and 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.

A selected list of annotated references is included at the end of each chapter. These references are suggestions for the individual who is interested in reading more about a particular subject. Additional relevant articles can be found in the suggested readings or in *The Clinical Practice of Echocardiography*. An online medical reference database is the best way to obtain more recent publications and to obtain a comprehensive list of all journal articles on a specific topic.

A special feature of this book that grew out of my experience teaching fellows and sonographers is The Echo Exam section at the end of the book. This section serves as a summary of the important concepts in each chapter and provides examples of the quantitative calculations used in the day-to-day clinical practice of echocardiography. The information in The Echo Exam is arranged as lists, tables, and figures for clarity. My hope is that The Echo Exam will also

serve as a quick reference guide when a review is needed and in daily practice in the echocardiography laboratory.

In the fourth edition, the text of all the chapters has been revised to reflect recent advances in the field, the suggested readings have been updated, and the majority of the figures have been replaced with recent examples that more clearly illustrate the disease process. In the first chapter, the sections on ultrasound physics have been expanded to provide the knowledge base needed for certification examinations in echocardiography. Many sections in other chapters have been extensively revised, including diastolic dysfunction, advanced echocardiographic modalities, cardiomyopathies, and adult congenial heart disease. The use of transesophageal imaging is explicitly integrated into each chapter. Updated guidelines for echocardiography have been included in each chapter when available. A new chapter on "Intraoperative Transesophageal Echocardiography" has been added to provide an introduction to this clinical application and to highlight some of the unique aspects of echocardiographic evaluation of patients undergoing surgical or percutaneous intervention. This chapter includes details of echocardiographic evaluation of patients undergoing mitral valve repair, acute and chronic aortic disease, and management of intracardiac instrumentation, such as left ventricular assist devices.

It should be emphasized that this textbook is only a starting point or frame of reference for learning echocardiography. Appropriate training in echocardiography 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. Further, echocardiography continues to evolve so that as new techniques, such as 3D echocardiography, become practical and widely available, practitioners will need to update their knowledge. Obviously, a textbook cannot replace the experience gained in performing studies on patients with a range of disease processes, and still photographs do not replace the need for acquisition and review of real-time data. Clearly defined guidelines for training in echocardiography have been published, as referenced in Chapter 5, that serve as guidelines for determining clinical competency in this technique. Although this textbook is not a substitute for appropriate training and experience, I hope 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. My appreciation extends to the many readers who provided suggestions for improvement; comments from readers are always welcome. The cardiac sonographers at the University of Washington deserve special thanks 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 David Diedrick, RDCS; Pam Clark, RDCS; Sarah Curtis, RDCS; Caryn D'Jang, RDCS; Merrit Foley, RDCS; Michelle Fujioka, RDCS; Carol Kraft, RDCS; Yelena Kovolenko, RDCS; Amy Loscher, RDCS; Chris McKenzie, RDCS; Joanna Sangco; Becky Schwaegler, RDCS; Erin Trent, RDCS; and Todd Zwink, RDCS.

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Finally, many thanks to my editor, Natasha Andjelkovic, at Elsevier for providing the support needed to write this edition, and to Anne Snyder and the production team for all the detail-oriented hard work that went into making this book a reality.

# GLOSSARY Abbreviations Used in Figures, Tables, and Equations

2D = two-dimensional
3D = three-dimensional
A-long = apical long-axis
A-mode = amplitude mode (amplitude versus depth)
$A$ = late diastolic ventricular filling velocity with atrial contraction
$A'$ = diastolic tissue Doppler velocity with atrial contraction
A2C = apical two-chamber
A4C = apical four-chamber
$AcT$ = acceleration time
AF = atrial fibrillation
AMVL = anterior mitral valve leaflet
ant = anterior
Ao = aortic or aorta
AR = aortic regurgitation
AS = aortic stenosis
ASD = atrial septal defect
ATVL = anterior tricuspid valve leaflet
AV = atrioventricular
AVA = aortic valve area
AVR = aortic valve replacement

BAV = bicuspid aortic valve
BP = blood pressure
BSA = body surface area

$c$ = propagation velocity of sound in tissue
CAD = coronary artery disease
cath = cardiac catheterization
$C_m$ = specific heat of tissue
cm/s = centimeters per second
cm = centimeters
CMR = cardiac magnetic resonance imaging
CO = cardiac output
cos = cosine
CS = coronary sinus
CSA = cross-sectional area
CT = computed tomography
CW = continuous-wave
Cx = circumflex coronary artery

$D$ = diameter
DA = descending aorta
dB = decibels
$dP/dt$ = rate of change in pressure over time
$dT/dt$ = rate of increase in temperature over time
dyne $\cdot$ s $\cdot$ cm$^{-5}$ = units of resistance

$E$ = early-diastolic peak velocity
$E'$ = early-diastolic tissue Doppler velocity

ECG = electrocardiogram
echo = echocardiography
ED = end-diastole
EDD = end-diastolic dimension
EDV = end-diastolic volume
EF = ejection fraction
endo = endocardium
epi = epicardium
EPSS = E-point septal separation
EROA = effective regurgitant orifice area
ES = end-systole
ESD = end-systolic dimension
ESV = end-systolic volume
ETT = exercise treadmill test

$\Delta f$ = frequency shift
$f$ = frequency
FL = false lumen
$F_n$ = near field
$F_o$ = resonance frequency
$F_s$ = scattered frequency
FSV = forward stroke volume
$F_T$ = transmitted frequency

HCM = hypertrophic cardiomyopathy
HOCM = hypertrophic obstructive cardiomyopathy
HPRF = high pulse repetition frequency
HR = heart rate
HV = hepatic vein
Hz = Hertz (cycles per second)

$I$ = intensity of ultrasound exposure
IAS = interatrial septum
ID = indicator dilution
inf = inferior
IV = intravenous
IVC = inferior vena cava
IVCT = isovolumic contraction time
IVRT = isovolumic relaxation time

kHz = kilohertz

$l$ = length
LA = left atrium
LAA = left atrial appendage
LAD = left anterior descending coronary artery
LAE = left atrial enlargement
lat = lateral
LCC = left coronary cusp
LMCA = left main coronary artery
LPA = left pulmonary artery

LSPV = left superior pulmonary vein
L-TGA = corrected transposition of the great arteries
LV = left ventricle
LV-EDP = left ventricular end-diastolic pressure
LVH = left ventricular hypertrophy
LVID = left ventricular internal dimension
LVOT = left ventricular outflow tract

M-mode = motion display (depth versus time)
MAC = mitral annular calcification
MI = myocardial infarction
MR = mitral regurgitation
MS = mitral stenosis
MV = mitral valve
MVA = mitral valve area
MVL = mitral valve leaflet
MVR = mitral valve replacement

$n$ = number of subjects
NBTE = nonbacterial thrombotic endocarditis
NCC = noncoronary cusp

$\Delta P$ = pressure gradient
$P$ = pressure
PA = pulmonary artery
PAP = pulmonary artery pressure
PCI = percutaneous coronary intervention
PDA = patent ductus arteriosus or posterior descending artery (depends on context)
PE = pericardial effusion
PEP = preejection period
PET = positron-emission tomography
PISA = proximal isovelocity surface area
PLAX = parasternal long-axis
PM = papillary muscle
PMVL = posterior mitral valve leaflet
post = posterior (or inferior-lateral) ventricular wall
PR = pulmonic regurgitation
PRF = pulse repetition frequency
PRFR = peak rapid filling rate
PS = pulmonic stenosis
PSAX = parasternal short-axis
PV = pulmonary vein
PVC = premature ventricular contraction
PVR = pulmonary vascular resistance
PWT = posterior wall thickness

$Q$ = volume flow rate
$Q_p$ = pulmonic volume flow rate
$Q_s$ = systemic volume flow rate

$r$ = correlation coefficient
$R$ = ventricular radius
$R_{FR}$ = regurgitant instantaneous flow rate
RA = right atrium
RAE = right atrial enlargement
RAO = right anterior oblique
RAP = right atrial pressure
RCA = right coronary artery
RCC = right coronary cusp
$R_e$ = Reynolds number
RF = regurgitant fraction
RJ = regurgitant jet
$R_o$ = radius of microbubble
ROA = regurgitant orifice area
RPA = right pulmonary artery
RSPV = right superior pulmonary vein
RSV = regurgitant stroke volume
RV = right ventricle
RVE = right ventricular enlargement
RVH = right ventricular hypertrophy
RVOT = right ventricular outflow tract

s = second
SAM = systolic anterior motion
SC = subcostal
SEE = standard error of the estimate
SPPA = spatial peak pulse average
SPTA = spatial peak temporal average
SSN = suprasternal notch
ST = septal thickness
STJ = sinotubular junction
STVL = septal tricuspid valve leaflet
SV = stroke volume or sample volume (depends on context)
SVC = superior vena cava

T½ = pressure half-time
TD = thermodilution
TEE = transesophageal echocardiography
TGA = transposition of the great arteries
TGC = time gain compensation
Th = wall thickness
TL = true lumen
TN = true negatives
TOF = tetralogy of Fallot
TP = true positives
TPV = time to peak velocity
TR = tricuspid regurgitation
TS = tricuspid stenosis
TSV = total stroke volume
TTE = transthoracic echocardiography
TV = tricuspid valve

$v$ = velocity
$V$ = volume or velocity (depends on context)
VAS = ventriculo-atrial septum
Veg = vegetation
$V_{max}$ = maximum velocity
VSD = ventricular septal defect
VTI = velocity-time integral

WPW = Wolff-Parkinson-White syndrome

$Z$ = acoustic impedance

| Symbols | Greek Name | Used for |
|---|---|---|
| $\alpha$ | alpha | Frequency |
| $\gamma$ | gamma | Viscosity |
| $\Delta$ | Delta | Difference |
| $\theta$ | theta | Angle |
| $\lambda$ | lambda | Wavelength |
| $\mu$ | mu | Micro- |
| $\pi$ | pi | Mathematical constant (approx. 3.14) |
| $\rho$ | rho | Tissue density |
| $\sigma$ | sigma | Wall stress |
| $\tau$ | tau | Time constant of ventricular relaxation |

## UNITS OF MEASURE

| Variable | Unit | Definition |
|---|---|---|
| Amplitude | dB | Decibels = a logarithmic scale describing the amplitude ("loudness") of the sound wave |
| Angle | degrees | Degree = $(\pi/180)$rad. Example: intercept angle |
| Area | $cm^2$ | Square centimeters. A two-dimensional measurement (e.g., end-systolic area) or a calculated value (e.g., continuity equation valve area) |
| Frequency (*f*) | Hz<br>kHz<br>MHz | Hertz (cycles per second)<br>Kilohertz = 1000 Hz<br>Megahertz = 1,000,000 Hz |
| Length | cm<br>mm | Centimeter (1/100 m)<br>Millimeter (1/1000 m or 1/10 cm) |
| Mass | g | Grams. Example: LV mass |
| Pressure | mmHg | Millimeters of mercury, 1 mmHg = 1333.2 dyne/$cm^2$, where dyne measures force in $cm \cdot g \cdot s^{-2}$ |
| Resistance | $dyne \cdot s \cdot cm^{-5}$ | Measure of vascular resistance |
| Time | s<br>ms<br>$\mu$s | Second<br>Millisecond (1/1000 s)<br>Microsecond |
| Ultrasound intensity | $W/cm^2$<br>$mW/cm^2$ | Where watt (W) = joule per second and joule = $m^2 \cdot kg \cdot s^{-2}$ (unit of energy) |
| Velocity (*v*) | m/s<br>cm/s | Meters per second<br>Centimeters per second |
| Velocity-time integral (VTI) | cm | Integral of the Doppler velocity curve (cm/s) over time (s), in units of cm |
| Volume | $cm^3$<br>mL<br>L | Cubic centimeters<br>Milliliter, 1 mL = 1 $cm^3$<br>Liter = 1000 mL |
| Volume flow rate (*Q*) | <br>L/min<br>mL/s | Rate of volume flow across a valve or in cardiac output<br>L/min = liters per minute<br>mL/s = milliliters per second |
| Wall stress | $dyne/cm^2$<br>$kdyn/cm^2$<br>kPa | Units of meridional or circumferential wall stress<br>Kilodynes per $cm^2$<br>Kilopascals where 1 kPa = 10 $kdyn/cm^2$ |

# KEY EQUATIONS

**Ultrasound Physics**

| | |
|---|---|
| Frequency | $f = \text{cycles/s} = \text{Hz}$ |
| Wavelength | $\lambda = c/f = 1.54/f(\text{MHz})$ |
| Doppler equation | $v = c \times \Delta f/[2F_T(\cos\theta)]$ |
| Bernoulli equation | $\Delta P = 4V^2$ |

**LV Imaging**

| | |
|---|---|
| Stroke volume | $SV = EDV - ESV$ |
| Ejection fraction | $EF(\%) = (SV/EDV) \times 100\%$ |
| Wall stress | $\sigma = PR/2Th$ |

**Doppler Ventricular Function**

| | |
|---|---|
| Stroke volume | $SV = CSA \times VTI$ |
| Rate of pressure rise | $dP/dt = 32$ mm Hg/time from 1 to 3 m/s of MR CW jet(sec) |

**Pulmonary Pressures and Resistance**

| | |
|---|---|
| Pulmonary systolic pressure | $PAP_{systolic} = 4(V_{TR})2 + RAP$ |
| PAP (when PS is present) | $PAP_{systolic} = 4(V_{TR})2 + RAP] - \Delta P_{RV-PA}$ |
| Pulmonary vascular resistance | $PVR \cong 10\,(V_{TR})/VTI_{RVOT}$ |

**Aortic Stenosis**

| | |
|---|---|
| Maximum pressure gradient (integrate over ejection period for mean gradient) | $\Delta P_{max} = 4\,(V_{max})^2$ |
| Continuity equation valve area | $AVA(cm^2) = [\pi(LVOT_D/2)^2 \times VTI_{LVOT}]/VTI_{AS-Jet}$ |
| Simplified continuity equation | $AVA(cm^2) = [\pi(LVOT_D/2)^2 \times V_{LVOT}]/V_{AS-Jet}$ |
| Velocity ratio | Velocity ratio $= V_{LVOT}/V_{AS-Jet}$ |

**Mitral Stenosis**

| | |
|---|---|
| Pressure half time valve area | $MVA_{Doppler} = 220/T^1/_2$ |

**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 | $RV = TSV - FSV$ |
| Regurgitant orifice area | $ROA = RSV/VTI_{AR}$ |

**Mitral Regurgitation**

| | |
|---|---|
| Total stroke volume OR 2D LV stroke volume | $TSV = SV_{MA} = (CSA_{MA} \times VTI_{MA})$ |
| Forward stroke volume | $FSV = SV_{LVOT} = (CSA_{LVOT} \times VTI_{LVOT})$ |
| Regurgitant volume | $RV = TSV - FSV$ |
| 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 (age 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**

$Qp:Qs = [CSA_{PA} \times VTI_{PA}]/[CSA_{LVOT} \times VTI_{LVOT}]$

# CONTENTS

# 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 of this technique and, more important, its limitations 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 might undergo needless, expensive, and potentially risky other 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. Since 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 refractions and compressions of any physical medium through which they pass (Fig. 1–1). Like other waves, sound waves are described in terms of (Table 1–1):

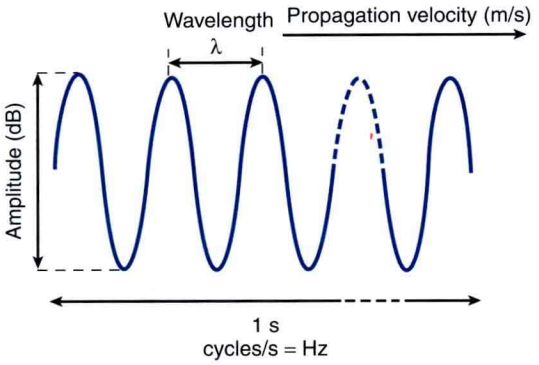


**Figure 1–1** Schematic diagram of an ultrasound wave.

- Frequency: cycles per second (cycles/s), 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.

*Wavelength* 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).

$$\lambda = c/f \qquad (1\text{–}1)$$

Since the propagation velocity in the heart is constant at 1540 m/s, using units of MHz for transducer frequency and dividing by 1000 to convert m to mm, the wavelength for any transducer frequency can be calculated as

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

**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 about 1540 m/s. | The velocity of propagation is similar in various soft tissues (blood, myocardium, liver, fat, etc.) but is much lower in lung and much higher in bone. |
| Wavelength ($\lambda$) | The distance between ultrasound waves: $\lambda = c/f = 1.54/f$ (in 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 decibels. On the decibel 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. |

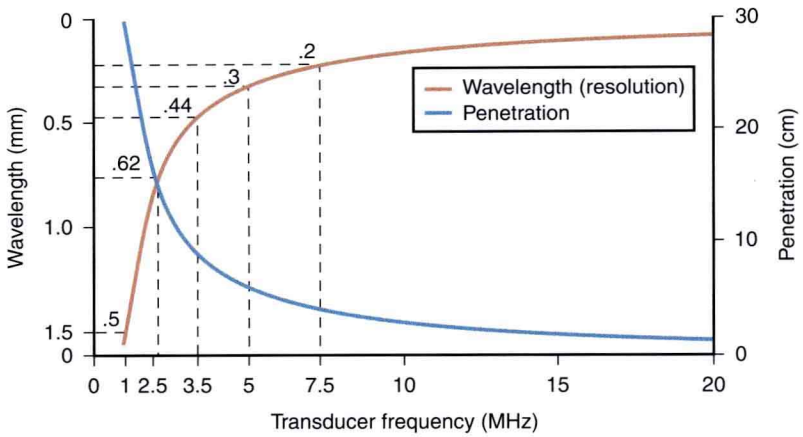


**Figure 1–2** Graph of transducer frequency (*horizontal axis*) versus wavelength and penetration of the ultrasound signal in soft tissue. Wavelength has been plotted inversely to show that resolution increases with increasing transducer frequency while penetration decreases. The specific wavelengths for transducer frequencies of 1, 2.5, 3.5, 5, and 7.5 MHz are shown.

as shown in Figure 1–2. For example, the wavelength emitted by a 5-MHz transducer can be calculated as

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

or as:

$$\lambda = 1.54/\text{s} = 0.308 \text{ mm}$$

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 tradeoff 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:

$$\text{Intensity } (I) = \text{power}^2 \qquad (1\text{–}2)$$

This relationship shows that if ultrasound power is doubled, intensity is quadrupled. Instead of using direct measures of pressure energy, ultrasound amplitude is described relative to a reference value using the decibel scale. Decibels are familiar to all of us as the standard description of the loudness of a sound. Decibels (dB) are logarithmic units based on a ratio of the measured amplitude ($A_2$) to a reference amplitude ($A_1$) such that

$$\text{dB} = 20 \log(A_2/A_1) \qquad (1\text{–}3)$$

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). 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. Either of these decibel scales may be 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 may be an option.

## 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 ($\mathcal{Z}$) depends on tissue density ($\rho$) and on the propagation velocity in that tissue ($c$):

$$\mathcal{Z} = \rho c \qquad (1\text{–}4)$$

Although the velocity of propagation differs between tissues (e.g., bone has a propagation velocity about twice as fast as blood), tissue density is the primary determinant of acoustic impedance for diagnostic ultrasound. Lung tissue has a very low density as compared with bone, which has a very high density. Soft tissues such

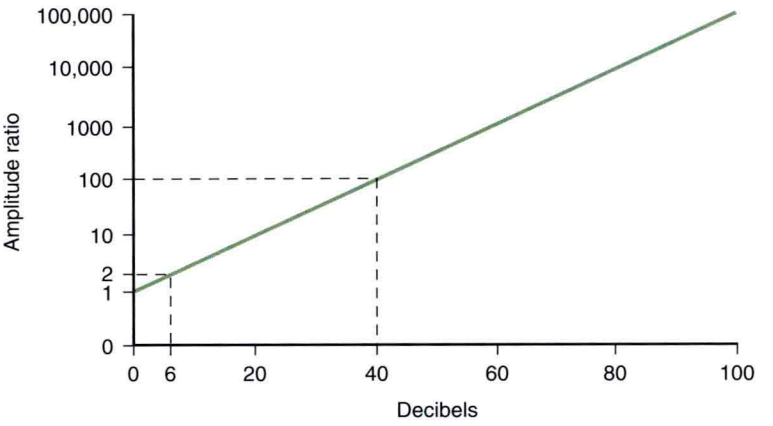


**Figure 1–3** Graph of the decibel scale (*horizontal axis*) showing the logarithmic relationship with the amplitude ratio (*vertical axis*). Note that a doubling or halving of the amplitude ratio corresponds to a 6-dB change, and a 100-fold difference in amplitude corresponds to a 20-dB change.

TABLE 1–2 Ultrasound-Tissue Interaction

| | Definition | Examples | Clinical Implications |
|---|---|---|---|
| Acoustic impedance ($Z$) | A characteristic of each tissue defined by tissue density ($\rho$) and propagation of velocity ($c$) as: $Z = \rho \times c$ | Lung has a low density and low propagation velocity, whereas bone has a high density and high 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 versus 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 when the ultrasound beam is perpendicular to the tissue interface. |
| Scattering | Radiation of ultrasound in multiple directions from small structures, such as 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 may be 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. |

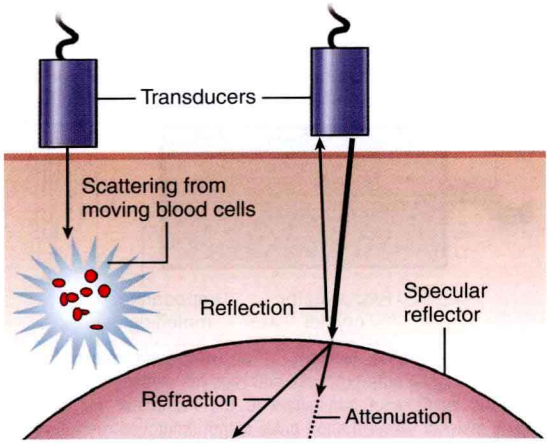


**Figure 1–4** Diagram of the interaction between ultrasound and body tissues. Doppler analysis is based on the scattering of ultrasound in all directions from moving blood cells with a resulting change in frequency of the ultrasound received at the transducer. 2D imaging is based on reflection of ultrasound 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.

as blood and myocardium have much smaller differences in 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

## 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 (1) the relative change in acoustic impedance between the two tissues and (2) the 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 angles of incidence and reflection are 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) images when too little or no reflected ultrasound reaches the transducer due to 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 may be radiated 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* that can be used to measure tissue motion 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, resulting in ultrasound artifacts, most notably a 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* due to 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 ($\alpha$) for each tissue is related to 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:

$$I_2 = I_1 \cdot e^{-2\alpha l} \qquad (1-5)$$

The attenuation coefficient for air is very high (about 1000×) compared with soft tissue, so that any air between the transducer and the cardiac structures of interest causes 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) attenuation is avoided by maintaining close contact between the transducer and the 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 due to 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 may cause thermal and mechanical bioeffects as discussed in "Bioeffects and Safety" below.

Overall attenuation is also 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 (such as quartz or a titanate ceramic) with the property that an applied electric current results in alignment of polarized 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.

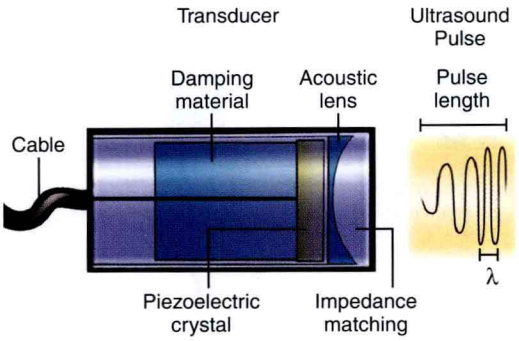


**Figure 1–5** Schematic diagram of an ultrasound transducer. The piezoelectric crystal both produces and receives ultrasound signals, with the electric input/output transmitted to the instrument via the cable. Damping material allows a short pulse length (improved resolution). The shape of the piezoelectric crystal, an acoustic lens, or electronic focusing (with a phased-array transducer) are used to modify the 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).

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, since 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, since a shorter time is needed for the same number of cycles at higher frequencies.

The range of frequencies contained in the pulse is described as its *frequency bandwidth*. A wider bandwidth