CLINICAL MANUAL AND REVIEW OF TRANSESOPHAGEAL ECHOCARDIOGRAPHY

THIRD EDITION

Joseph P. Mathew
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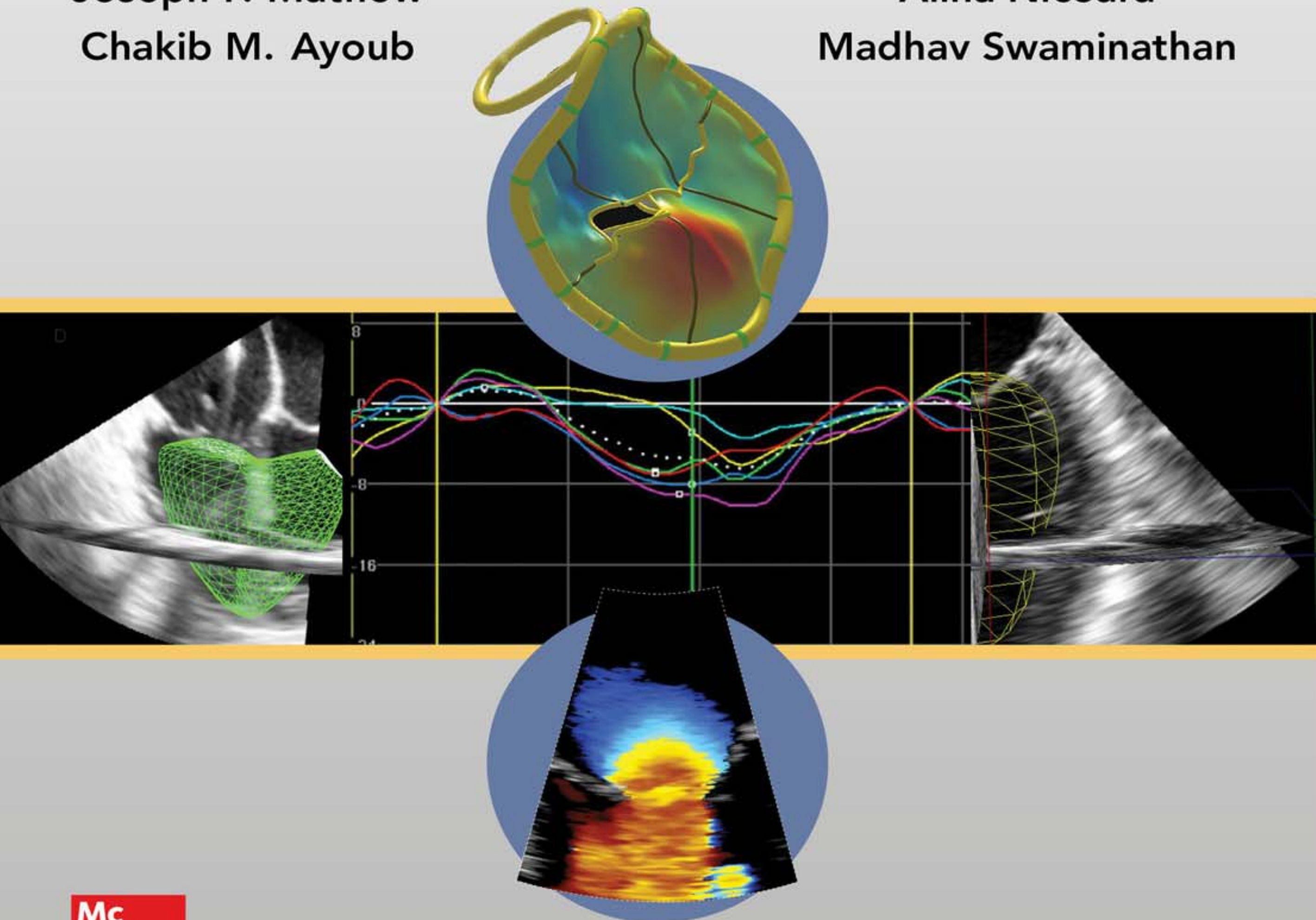
Madhav Swaminathan



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CLINICAL MANUAL AND REVIEW OF TRANSESOPHAGEAL ECHOCARDIOGRAPHY

Third Edition

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To my mentors, Drs. Paul Barash, Jerry Reves, and Mark Newman. How blessed I have been to have you all as role models and to learn from such great leaders! Thank you for your commitment, sacrifice, and legacy and most importantly, for instilling in me a "vision for what can be" as well as the "passion for what should be".

Joseph P. Mathew

To my parents, Rodica and Ioan.

Alina Nicoara

I dedicate this book to the many great people who have unselfishly given their time and talent to mentor and guide others. To my mentors, teachers and friends, you have been an integral part of my career. Working with you has always been a motivating and memorable journey, which has guided me so far, and always will. I hope one day to inspire others as you have inspired me.

Chakib M. Ayoub

To Ratna, my wife, best friend, and confidante, for her unconditional support.

My son, Raghav, for always inspiring me.

My son, Abhinav, for his wit, humor, and sage advice often beyond his years.

Madhav Swaminathan

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Foreword

Over the past 60 years, echocardiography has become the most important and widely used imaging modality in cardiovascular medicine. And even though the core technology still relies on the same fundamental physical properties of reflected waves, the variety of innovations and applications that have evolved to enable sophisticated imaging of cardiac structure and function in three-dimensional space is astonishing. The underlying principles are simple enough: ultrasound waves are sent from a transducer, then the reflected waves are analyzed in two domains: a time-intensity domain (to characterize the structure) and a frequency-shift domain (to assess the speed of motion). With modern transducer technology and computer processing speeds, echo images have greater spatial and temporal resolution than ever, and the usefulness of echo across all forms of cardiovascular disease is without parallel. One cannot be a cardiovascular specialist without an in-depth knowledge of echocardiography.

This volume goes a long way to addressing this need. While transthoracic echocardiography is the initial approach for the majority of clinical conditions, transesophageal echocardiography (TEE) remains essential since its introduction in the 1980s, as it provides images with much higher resolution, particularly in the cardiac base and in technically difficult studies. This is crucial in the care of the critically ill patient or during cardiac interventions, where diagnostic accuracy is paramount. Drs. Joseph P. Mathew, Alina Nicoara, Chakib M. Ayoub, and Madhav Swaminathan lay a solid foundation for understanding the intricacies of transesophageal echocardiography, from the basic principles of ultrasound to various clinical applications of TEE, including its role in the critically ill, in monitoring heart function in the operative theater, and in evaluating the immediate results of cardiac procedures.

This third edition of *Clinical Manual and Review of Transesophageal Echocardiography* accomplishes the difficult task of speaking with ease to both the novice and the expert, while updating the text of their second edition to include recent improvements in the field, inclusive of strain imaging. The authors cover all the novel uses of three-dimensional imaging with a focus on those that apply to valvular heart surgery and the critical care setting, offering wisdom accumulated through long experience. The illustrations and instructional videos included with this volume reflect the authors' deep expertise in both the technique and the teaching of TEE; questions at the end of chapters and the TEE practice exams will help prepare those who intend on taking certification examinations. This third edition also incorporates a superb online supplement with instructional videos.

In short, this clinical manual and review of transesophageal echocardiography will enlighten the reader and promote the expert application of transesophageal echocardiography in the care of high-risk patients.

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Preface

The third edition of the *Clinical Manual and Review of Transesophageal Echocardiography* is intended to be an indispensable resource and the standard reference manual in the field of transesophageal echocardiography (TEE). Completely updated, reorganized, and expanded, this edition has been redesigned to offer anesthesiologists, cardiologists, cardiothoracic surgeons, emergency room physicians, intensivists, and sonographers a concise yet comprehensive coverage of the key principles, concepts, and current practices in TEE.

Since the publication of the first edition in 2005, and the second edition in 2010, the field has witnessed continuous growth at a rapid pace. Preparing the third edition was a complex undertaking, as we attempted to balance content, format, style, integration, and innovation, while recognizing the need to stay in the zone between excessively complex and over simplified.

This edition features a sectional format, each containing chapters that were reviewed and updated to provide a comprehensive discussion of the physiology, pathophysiology, and echocardiographic approach for normal and common disease states. Whenever possible, important clinical information has been aligned with the principles of cardiovascular physiology, and echocardiographic techniques. In addition, narrative text, charts, and graphs have been effectively integrated to provide rapid access to key clinical information for the purpose of improving clinical management. With a dedicated section highlighting the practice exam along the numerous multiple-choice questions after almost every chapter, in addition to the online instructional videos provided to our readers, this edition will serve as an excellent source of current clinical information on TEE for trainees and more experienced anesthesiologists preparing for board examinations in both basic and advanced perioperative echocardiography.

In addition to several distinguished new authors, we welcome Dr. Alina Nicoara as a co-editor. We have been privileged to collaborate with an outstanding group of colleagues that are prominent experts in their fields. We are grateful and acknowledge their hard work, dedication, selfless commitment and valued contributions in this collective responsibility. It is their excellence, attention to detail, passion for echocardiography, and vast knowledge and experience that allowed this project to proceed smoothly. We are also thankful to the many readers of the first and second editions who offered words of encouragement and even advice on how the book could be improved—many of those suggestions have been incorporated into the current edition. Despite the changes, however, we hope that we have retained the elements that made the first two editions successful.

Finally, we once again recognize and are indebted to our mentors—those who instilled in us the passion for echocardiography. We gratefully acknowledge the contributions of Drs. Dinesh Kurian, Martin Sigurdsson, and Nathan Waldron in reviewing all of the questions and answers in the book. Our sincere appreciation also goes to our assistants, Melinda Macalino, Jaime Cooke, and Rabih Mukalled, for their dedication, enthusiasm, and patience. In addition, we would like to thank the staff at McGraw-Hill including Brian Belval, Andrew Moyer, Jason Malley and Christie Naglieri for their continued support with this project.

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Fundamentals of Echocardiography 1

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Physics of Two-Dimensional and Doppler Imaging

1

Brian P. Barrick, Mihai V. Podgoreanu, and Edward K. Prokop

BASICS OF ULTRASOUND[1–3]

Nature and Properties of Ultrasound Waves

Humans can hear sound waves with frequencies between 20 and 20,000 hertz. Frequencies higher than this range are called ultrasound (US). A *sound wave* can be described as a mechanical, longitudinal wave composed of cyclic compressions and rarefactions of molecules in a medium. This is in contrast to electromagnetic waves, which do not require a medium for propagation. Three **acoustic variables** identify sound waves:

- **Pressure:** force within an area (measured in pascals)
- **Density:** mass within a volume (measured in kg/cm^3)
- **Distance:** motion measured in length (e.g., millimeters, centimeters)

Three parameters can be used to describe the absolute and relative strength ("loudness") of a sound wave:

- **Amplitude:** The amount of change in one of the previously mentioned acoustic variables. Amplitude is equal to the difference between the average and the maximum values of an acoustic variable (or half the "peak-to-peak" amplitude).
- **Power:** The rate of energy transfer, expressed in watts (joules/second). Power is proportional to the square of the amplitude.
- **Intensity:** The energy per unit cross-sectional area in a sound beam, expressed in watts per square centimeter (W/cm^2). This is the parameter used most frequently when describing the biological safety of US.

The operator *can* modify the three parameters described, but it should be noted that modifying these parameters is not the same as adjusting receiver gain, which is a postprocessing function.

Changes (usually in intensity) can also be expressed in a relative, logarithmic scale known as **decibels (dB)**. In common practice, the lowest-intensity audible sound (10^{-12} W/M^2) is assigned the value of 0 dB. An increase of 3 dB represents a two-fold increase in intensity, and an increase of 10 dB represents a ten-fold increase in intensity. This means that a sound with an intensity of 120 dB is 1 trillion (10^{12}) times as intense as a sound of 0 dB.

Four additional parameters that are inherent to the sound generator (transducer) and/or the medium through which the sound propagates are also used. When referring to a single transducer (piezoelectric) element in a pulsed US system, these *cannot* be manipulated by the operator:

- **Period:** The duration of a single cycle. Typical values for clinical US range from 0.1 to 0.5 microseconds (μs).
- **Frequency (*f*):** The number of cycles per unit time. One cycle per second is one hertz (Hz). US is defined as a sound wave with a frequency greater than 20,000 Hz. Values that are relevant in clinical imaging modalities such as echocardiography and vascular ultrasound range from 2 to 15 megahertz (MHz).

Period and frequency are reciprocals. Period $= 1/f$.

- **Wavelength (λ):** The distance traveled by sound in one cycle (0.1 to 0.8 mm).

  Wavelength and frequency are inversely proportional and are related by propagation speed through the formula $\lambda = c/f$.

- **Propagation speed (*c*):** The speed of sound in a medium, determined by the characteristics of the medium through which it propagates. Propagation speed does *not* depend on the amplitude or frequency of the sound wave. It is directly proportional to the stiffness and inversely proportional to the density of the medium.

Sound propagates at 1540 m/s for average human soft tissue, including heart muscle, blood, and valve tissue. Other useful values are 330 m/s for air and 4080 m/s for skull bone.

Properties of Pulsed Ultrasound

Continuous waves are not useful for structural imaging. Instead, US systems use brief pulses of acoustic

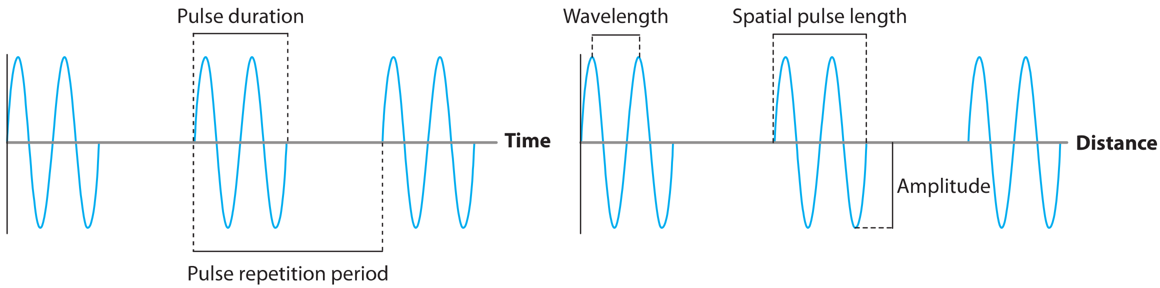

FIGURE 1–1. Physical parameters describing continuous and pulsed ultrasound waves.

signal. These are emitted from the transducer during the "on" time and are received during the "off" time. One pulse typically consists of three to five cycles.

Pulsed US can be described by five parameters (Fig. 1-1):

- **Pulse duration:** The time a pulse is "on," which is very short (0.5 to 3 μs).
- **Pulse repetition period:** The time from the start of one pulse to the start of the next pulse, which includes the listening time. Typical values are 0.1 to 1 ms.
- **Spatial pulse length:** The distance from the start to the end of a pulse (0.1 to 1 mm).
- **Duty factor:** The percentage of time the transducer is actively transmitting US, usually 0.1% to 1%. This means that the transducer element acts as a receiver over 99% of the time.
- **Pulse repetition frequency (PRF):** The number of pulses that occur in 1 second, expressed in Hz. PRF is reciprocal to pulse repetition period. Typical values are 1000 to 10,000 Hz (not to be confused with the frequency of the US within a pulse, which is many times greater).

PRF is inversely proportional to imaging depth. Because sound takes time to propagate, a deeper image requires more listening time. Therefore, with a deeper image, the transducer can emit fewer pulses per second. This concept will also be important for the discussion of Doppler ultrasound.

The relation between the depth of a reflector and the time it takes for an US pulse to travel from the transducer to the reflector and back to the transducer (time-of-flight) is called the *range equation*:

$$\text{distance to reflector (mm)} = \text{propagation speed (mm/}\propto\text{s)} \cdot \text{time-of-flight (}\propto\text{s)/2}$$

This allows the US system to calculate the distance to a certain structure by measuring only the time-of-flight. Assuming that soft tissue has a uniform propagation speed of 1540 m/s, or 1.54 mm/μs, this means that time-of-flight increases by 13 μs for every 1 cm of depth of the reflector. This value is important for imaging and for Doppler US.

Propagation of Ultrasound Through Tissues

The most important effect of a medium on the US wave is *attenuation*—the gradual decrease in intensity (measured in dB) of an US wave. Attenuation results from three processes:

- **Absorption:** conversion of sound energy to heat energy.
- **Scattering:** diffuse spread of sound from a border with small irregularities.
- **Reflection:** return of sound to the transducer from a relatively smooth border between two media. It is reflection that is important for imaging.

Different tissues attenuate by different processes and at different rates:

- Air bubbles reflect much of the US that engages them and appear very echo dense (bright). Because sound attenuates the most in air, information distal to an air bubble is often lost as a result.
- Lung, being mostly air filled, causes much scatter and results in the most attenuation of US by tissue.
- Bone absorbs and reflects US, resulting in somewhat less attenuation than lung.
- Soft tissue and blood attenuate even less than bone.
- Water attenuates sound very little, mostly by absorption, with very little reflection. It is therefore very echo lucent (appears black on images).

Within soft tissue, attenuation is proportional to both the US frequency and path length, and can be expressed by the following equation:

$$\text{Attenuation (dB)} = 0.5\ \text{dB/(cm} \cdot \text{MHz)} \cdot \text{path length (cm)} \cdot \text{frequency (MHz)}$$

Therefore, one may conclude that high-frequency US has greater attenuation, has poorer penetration, and is less effective at imaging deeper structures.

Less than 1% of the incident US is usually reflected at the boundary between different soft tissues. The interfaces between air and tissue and between bone and tissue are strong reflectors and can result in several types of artifacts (see Chapter 6).

As the US beam strikes a boundary between two media, three phenomena may occur:

- Reflection: can be further broken down into *specular reflection* and *diffuse reflection* or *backscatter*
- Transmission
- Refraction

Reflection of the transmitted US signal is the basis of US imaging. It can occur only if there is a difference in the *acoustic impedance* (measured in MRayls) between the two media, and is dependent on the angle of incidence of the US beam at the interface. Acoustic impedance is a property of the media, not of the US beam. It is directly proportional to both density and propagation speed of the material.

Specular reflectors have large, smooth surfaces, or have irregularities that are larger than the wavelength of the US beam. They are angle dependent, reflecting US best at normal incidence (90 degrees, or perpendicular to the boundary).

Scatter reflectors (the "signal" used in US imaging) have irregularities that are about the same size or smaller than the wavelength of US that strikes the boundary. Scatter reflectors are also not angle dependent. A special type of scattering is termed *Rayleigh scattering,* and this occurs when US strikes an object much smaller than the beam's wavelength (such as a red blood cell). Sound is scattered uniformly in all directions.

Refraction is a process associated with transmission and refers to the change of wave direction upon crossing the interface between two media. Refraction can occur only when the propagation speeds in the two media are different and the incident angle is oblique (Fig. 1-2). Refraction is described by *Snell's law*:

sine (refracted angle)/sine (incident angle) = speed of sound in medium 2/speed of sound in medium 1

Thus, if the speed of sound in medium 2 is less than the speed of sound in medium 1, then the transmission angle is less than the incident angle. Similarly, if the speed of sound in medium 2 is greater than the speed of sound in medium 1, then the transmission angle is greater than the incident angle.

Because it violates the assumption that US travels in a straight line, refraction may result in image artifacts (e.g., a second copy of a true reflector).

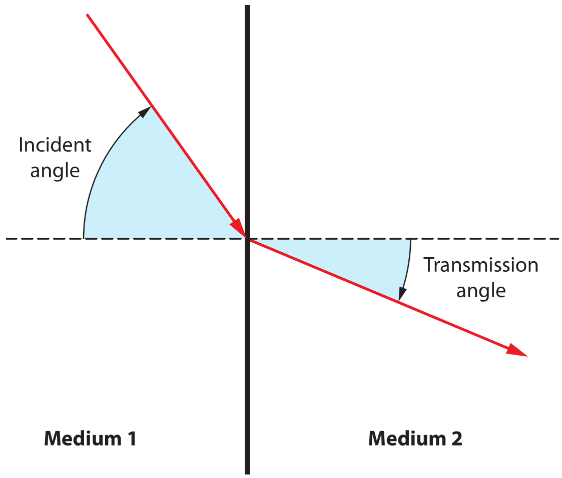

FIGURE 1–2. An illustration demonstrating refraction. In this example, the propagation speed of medium 1 is greater than medium 2, resulting in a lower transmission angle.

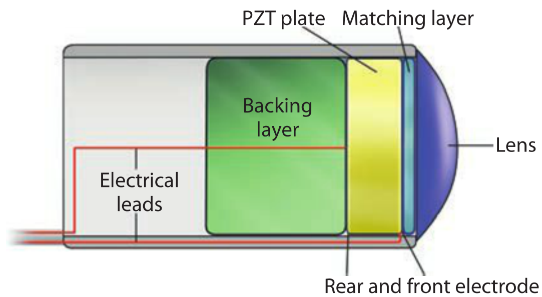

FIGURE 1–3. Anatomy of an ultrasound transducer.

ULTRASOUND TRANSDUCERS

Simply put, an ultrasound transducer (Fig. 1-3) is a device that converts electrical energy into high-frequency acoustic energy, and vice versa. US transducers contain crystals that change shape when an electrical potential is applied (*reverse piezoelectric effect*), as during sound transmission, and also create voltage when mechanically deformed (*piezoelectric effect*), as during sound reception. The most common crystals in US systems are composed of lead, zirconate, and titanate (PZT). The frequency of the US generated by each piezoelectric element is related to the thickness and the propagation speed of the crystal by the formula:

Frequency (MHz) = the material's propagation speed (mm/∝s) / twice the thickness (mm)

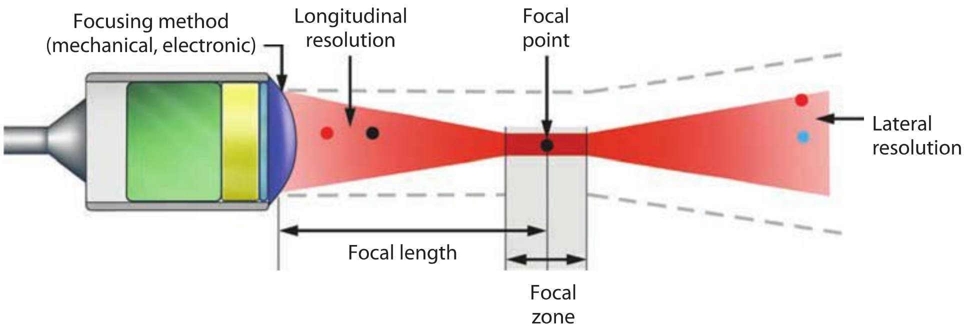

FIGURE 1–4. Anatomy of an ultrasound beam.

In addition to the crystal, there is a backing (damping) material with a high characteristic acoustic impedance and ability to absorb US that is designed to reduce unwanted ringing of the crystal. This leads to a shorter pulse length and improves resolution of the picture. The backing layer also increases the range of frequencies (or bandwidth) around the resonant frequency of the crystal. A wide bandwidth in an imaging transducer is useful because it gives the operator a limited ability to adjust the frequency of the US beam, thereby optimizing imaging. Frequencies used in transesophageal echocardiography (TEE) typically range from 2.0 to 7.0 MHz.

There is also a matching layer in front of the crystal. This layer is designed to have an acoustic impedance between that of the transducer material and the soft tissue it contacts, thus increasing transmission of US. The ideal matching layer has a thickness of one-quarter of the wavelength.

The sound beam produced by a single crystal whose thickness is one-half the wavelength of emitted sound spreads in a hemispherical pattern. The beam emitted by an US transducer composed of several crystals, however, has a characteristic hourglass shape due to constructive and destructive interference of the wavelets from each crystal. This is referred to as *Huygens' principle*. The *focal point* or *focus* is the location where the beam reaches its maximal intensity and minimum diameter (Fig. 1-4). Here the beam is about half the width of the transducer. The near area, or area between the transducer and focus, is called the Fresnel zone. The far area after the focus is called the Fraunhofer zone.

The simplest transducer can be composed of a single piezoelectric crystal that produces a 2D image via *mechanical scanning*. More commonly, multiple elements are arranged in arrays. In *linear switched arrays,* the simplest type of array, the elements are arranged in a line and fire simultaneously. In *phased arrays* (linear, annular, or convex), the elements fire with very small time delays, on the order of 10 nanoseconds. Phased arrays allow for electronic focusing and steering of the US beam.

If all of the elements are fired simultaneously, as in a linear switched array, the image would be rectangular and the focus would be fixed. Changing the pattern of time delays in element firing, as in phased arrays, allows for steering of the beam, resulting in a wider scan area (sector shaped). It also allows for adjustment of the focal point.

Conventional 2D US transducers—both transthoracic and transesophageal—have 64 to 128 elements arranged in a single row. Current transducers have both full 2D and 3D capabilities, with fully sampled matrix array transducers containing a considerably larger number of imaging elements per row, for a total of approximately 3000 elements (Table 1-1).[4,5] Novel electronic circuitries, improved ultrasound crystal technology, and increased computer-processing power have led to 3D transducers with a small footprint and have facilitated the use of routine 3D imaging in patient care. This topic is developed in further detail in Chapter 23.

INSTRUMENTATION

Components of an Ultrasound System

Any US system has six components:

Transducer: Converts electrical energy into acoustic energy and vice versa.

Pulser: Controls the electrical signals sent to the transducer. Controls PRF, pulse amplitude, and pulse repetition period. It is also responsible for electronic steering and focusing in phased arrays.

Receiver: Processes returning signals to produce an image on a display. Processing occurs in the following order:

1. *Amplification*: overall gain, 50 to 100 dB.
2. *Compensation*: more specifically, time gain compensation. Adjusts for increased attenuation with depth.

Table 1–1. Summary of transducer properties.

Transducer Type	Image Shape	Steering Technique	Focusing Technique	Crystal Defect
Mechanical	Sector	Mechanical	Fixed	Image loss
Linear switched array	Rectangular	None	Fixed	Vertical line dropout
Linear phased array	Sector	Electronic	Electronic	Poor steering and focusing
Annular phased array	Sector	Mechanical	Electronic	Horizontal line dropout
Convex sequential array	Blunted sector	None	Fixed	Vertical line dropout
Convex phased array	Blunted sector	Electronic	Electronic	Poor steering and focusing
Vector array	Flat top sector	Electronic	Electronic	Poor steering and focusing
Matrix array	Sector	Electronic	Electronic	Poor steering and focusing

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3. *Compression*: reduces the dynamic range of the signals to match the dynamic range of the system's electrical components. Does not change the relative value of the returning signals.
4. *Demodulation*: makes the image more suitable for viewing.
   a. Rectification converts all returning signals into positive amplitude.
   b. Smoothing converts signal bursts into a single deflection for each reflector
5. *Rejection*: elimination of low-level signals.

Display: Formerly a cathode ray tube, now usually consists of a computer monitor screen.

Storage media: Archiving of data (optical disk, DVD, network server).

Master synchronizer: Integrates all the individual components of the system.

Ultrasound Imaging

The modes of displaying returning echoes are as follows:

A (amplitude) mode: No longer used in clinical echocardiography. Displays upward deflections with height proportional to the amplitude of the returning echo and location proportional to the depth of the reflector (x-axis: reflector depth; y-axis: amplitude of echo). This mode only displays one scan line.

B (brightness) mode: Displays spots with brightness proportional to the amplitude of the echo and location proportional to the depth of the reflector (x-axis: reflector depth; z-axis: amplitude of echoes; there is no y-axis). B mode echocardiography can be further classified as follows:

- **M (motion) mode:** A continuous B-mode display. Displays one scan line versus time. Allows for a high frame rate, accuracy of linear measurements, and tracking of motion of reflectors (x-axis: time; y-axis: reflector depth).
- **Two-dimensional imaging** is a line of B-mode echo data moved in an arc through a section of tissue in a back-and-forth fashion. This can be achieved with mechanical or electronic steering of the B-mode echo beam. Images are generated as a series of frames displayed in rapid fashion to produce the impression of constant motion.
- **Three-dimensional imaging** displays pyramidal datasets consisting of volume elements or voxels. The 3D datasets appear three-dimensional on the 2D display monitors by creating the perception of depth through a range of colors and opacities.

Determinants of Two-Dimensional Resolution

The ability of an US system to image accurately is termed *resolution*. *Spatial resolution* is defined as the minimum separation between two reflectors where they can still be identified as different structures. Spatial resolution has been described in terms of distinguishing structures parallel to the US beam (***longitudinal*** *or* ***axial*** *resolution*) or perpendicular to the US beam (***lateral*** *resolution*).

Synonyms for longitudinal resolution include *axial, radial, range,* and *depth* (LARRD). Synonyms for lateral resolution include *angular, transverse,* and *azimuth* (LATA).

Longitudinal resolution = spatial pulse length/2. Therefore, longitudinal resolution can be improved by shortening the spatial pulse length. Given the same number of cycles per pulse, higher-frequency US will result in a shorter pulse length. Longitudinal resolution is typically better than lateral resolution.

Lateral resolution is approximately equal to the US beam diameter, and it is best at the focus point where the beam is the narrowest. The distance from the transducer to the focus point represents the focal length. Focal length is directly proportional with the frequency and the transducer diameter. Therefore,

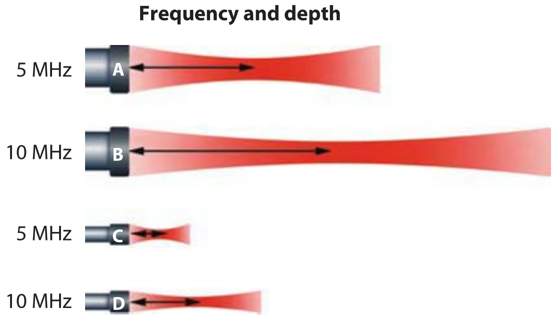

FIGURE 1–5. Transducer size, frequency, and focal depth.

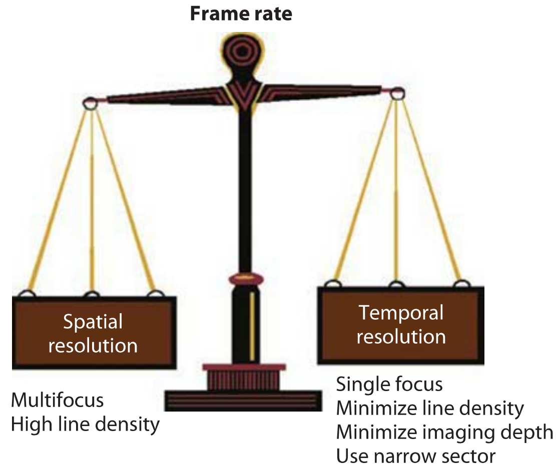

FIGURE 1–7. Relation between frame rate, spatial resolution, and temporal resolution. Improving temporal resolution is achieved at the expense of spatial resolution.

higher US frequency will result in a deeper area of focus (Fig. 1-5) and less divergence in the far field.

Note that both longitudinal and lateral resolutions are improved with high-frequency US. In choosing the settings of an US system, there is a tradeoff between the ability to obtain high-resolution images and the ability to image deeper structures (Fig. 1-6).

The ability to accurately locate moving structures at a given time is termed *temporal resolution*. Temporal resolution is proportional to the numbers of frames per second (*frame rate*). Factors that improve temporal resolution (by increasing the frame rate) are:

1. Minimizing imaging depth
2. Using single focus imaging (one pulse/line)

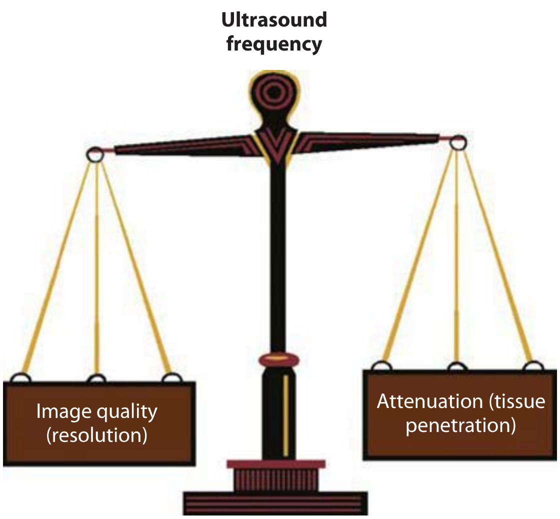

FIGURE 1–6. Relation between ultrasound frequency, image resolution, and tissue penetration. Image resolution improves at higher frequencies, but at the expense of tissue penetration.

3. Using a narrow sector
4. Minimizing line density

Because using multifocus imaging and high line density results in better lateral resolution, improved temporal resolution is achieved at the expense of spatial resolution (Fig. 1-7).

Harmonic Imaging

As ultrasound travels through tissue, it generates additional sound frequencies, which are multiple, or harmonics, of the transmitted frequency. The farther the US travels, the more harmonics it produces. By properly filtering the returning signal, which contains the harmonic frequencies, an US machine equipped for harmonic imaging can selectively display images created with harmonic energy.

The reception of the transmitted ultrasound at the second harmonic (double) frequency improves tissue visualization, especially in situations of poor-quality imaging with fundamental frequency, by enhancing both myocardial and valvular tissue. Normal structures may appear abnormally thickened; therefore, care should be taken on interpreting harmonic images. Harmonic imaging also increases the signal-to-noise ratio and limits the creation of artifacts, especially in the proximity of the transducer (near field)[6] (see Chapter 6).

PRINCIPLES OF DOPPLER ULTRASOUND

The *Doppler effect* is defined as the change in the frequency of sound emitted or reflected by a moving object. The amount of change is termed the *Doppler shift*. It is important to note that though both the transmitted and reflected frequency are ultrasonic (MHz range), the actual Doppler shift is in the audible range (20 to 20,000 Hz).

The most common applications of Doppler US are to measure velocity (magnitude and direction) of blood flow and, more recently, tissue. The *Doppler equation* is as follows:

$$\text{Doppler shift (expressed in Hz)} = (2 \cdot v \cdot Fi \quad \text{cosine } \theta)/c$$

- v = the velocity of the moving object
- Fi = the incident frequency, or frequency emitted by the transducer
- θ = the angle between the incident US beam and the direction of movement
- c = the propagation speed of US in the medium (a constant 1540 m/s in soft tissue)

If the object is moving directly toward ($\theta = 0°$) or away from ($\theta = 180°$) the transducer and v is expressed in units of m/s, then cosine θ is 1 and the equation simplifies to:

$$\text{Doppler shift} = (v \cdot Fi)/770$$

Because the Doppler shift varies with the cosine of the angle of beam incidence (θ), the maximum measurable velocity decreases as θ increases. When movement is perpendicular (90 degrees) to the beam, no Doppler shift is detected. Therefore, only measurements obtained with (θ) smaller than 20 degrees are considered accurate.

In practice, the machine *measures* a Doppler shift and *calculates* a velocity. It also assumes θ is 0 degrees or 90 degrees. Rearranging the simplified Doppler equation to reflect this gives us the following:

$$v = 770 \cdot (\text{Doppler shift}/Fi)$$

When reflected (backscatter) signals are received at the transducer, the difference between the transmitted and reflected frequency is determined, analyzed by fast Fourier transform, and then displayed on the screen as a Doppler envelope. This process is known as spectral analysis and results in a display of the following:

- Direction of blood flow: Flow toward the transducer results in an increased frequency (positive Doppler shift displayed above the baseline), whereas flow away from the transducer results in a decreased frequency (negative Doppler shift displayed below the baseline)
- Velocity of frequency shift
- Signal amplitude

Spectral Doppler (in contrast to color Doppler) can be further divided into pulsed wave and continuous wave.

Pulsed Wave Doppler

Pulsed wave Doppler uses one crystal that alternates between sending and receiving an US beam. A timed pulse allows sampling from a discrete area of about 1 to 3 mm, selected by the operator, known as the *sample volume*. This allows for *range discrimination* (Fig. 1-8). Because the same element acts as both sender and receiver, the transducer must wait for the pulse to complete a round trip before emitting another pulse. As an example, if the sampling volume is 5 cm from the probe, the transducer must wait 65 μs (10 cm/154,000 cm/sec) before sending the next pulse.

Because sampling is intermittent, the pulse repetition frequency limits the maximum Doppler shift (and thus maximum velocity) that can be measured accurately. Velocities higher than this maximum velocity will appear to wrap around on the display, a phenomenon known as *aliasing* (see Chapter 5). The Doppler frequency shift at which aliasing occurs, equal to PRF divided by 2, is termed the *Nyquist limit*.

For example, if a 5-MHz transducer can only send out about 15,000 pulses per second, the Nyquist limit is 7500 Hz (15,000/2). Using the velocity equation shown earlier, the maximum velocity that can be measured without aliasing is about 1.15 m/s (770 × (7500/5,000,000)).

Methods to avoid aliasing include:

1. Using continuous wave Doppler (described later)
2. Changing the view to bring the area of interest closer to the probe (shallower depth)
3. Using a transducer with a lower incident frequency (results in lower Doppler shift for given flow velocity; see the equation earlier)
4. Adjusting the scale to its maximum
5. Moving the baseline up or down (makes the picture "prettier" but does not eliminate aliasing)

From a practical standpoint, pulsed wave Doppler should be used when measuring relatively low-flow velocities (<~1.2 m/s) in specific areas of interest (e.g., pulmonary vein flow, mitral valve inflow).

Compared to imaging ultrasound, pulsed wave Doppler requires greater output power, longer pulse lengths, and a higher pulse repetition frequency.

When the velocity of the tissue becomes the object of measurement (Doppler tissue imaging), the system is set as a low-pass filter. This means that low-velocity, high-amplitude signals are preferentially displayed. Doppler tissue imaging is discussed in more detail in Chapter 5.

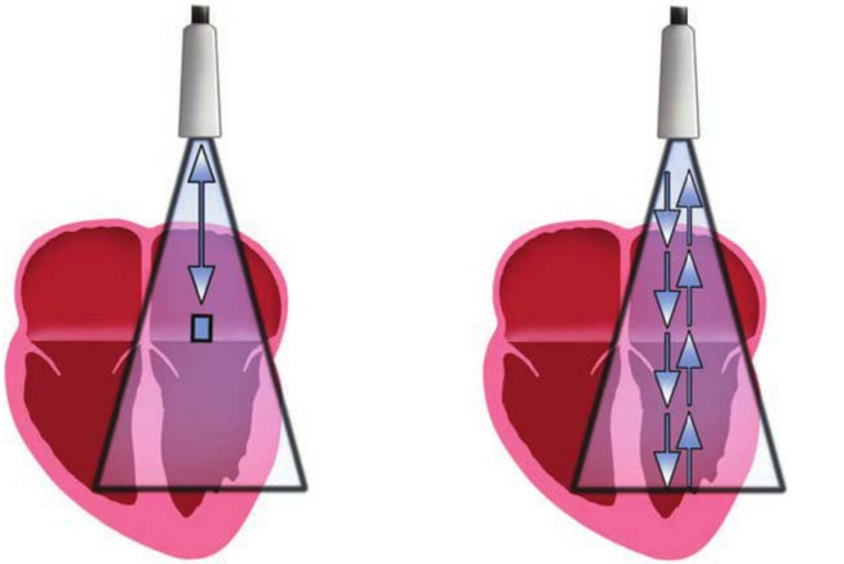

FIGURE 1–8. Characteristics of pulsed wave and continuous wave Doppler.

Continuous Wave Doppler

Continuous wave Doppler uses two crystals in the transducer: one to constantly send US waves and the other to continuously receive. The PRF can thus be extremely high. This continuous sampling allows determination of high-velocity flow. However, because echoes come from anywhere along the length of the beam, continuous sampling prevents determination of the location of maximum measured velocity, termed *range ambiguity* (see Fig. 1-8).

From a practical standpoint, continuous wave Doppler should be used when measuring velocities >~1.2 m/s. (e.g., regurgitant jets, stenotic valves).

Color Flow Doppler

Color flow Doppler is a pulsed US technique that color-codes Doppler information and superimposes it on a 2D image, providing information on the direction of flow and semiquantitative information on the mean velocities of flow. It has the characteristics of pulsed wave Doppler (range discrimination and aliasing). Color flow Doppler uses packets of multiple pulses (3 to 20 per scan line), and therefore has a low temporal resolution (Fig. 1-9). It then employs spectral analysis methods to estimate the mean velocity at each depth. The information on the direction

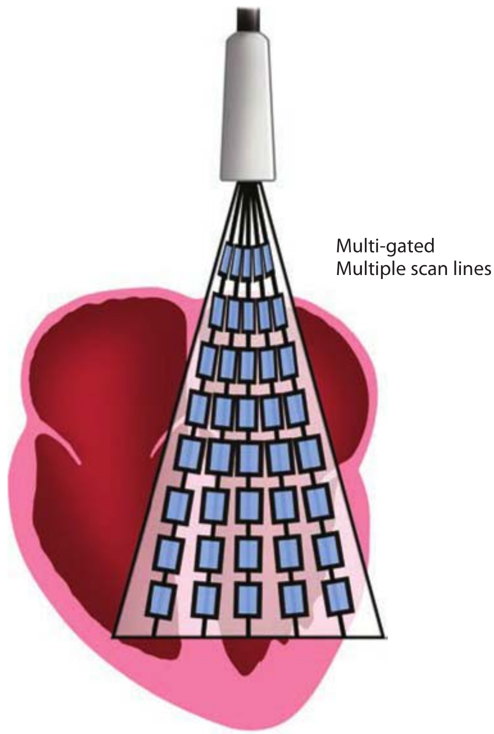

FIGURE 1–9. Characteristics of color flow Doppler.

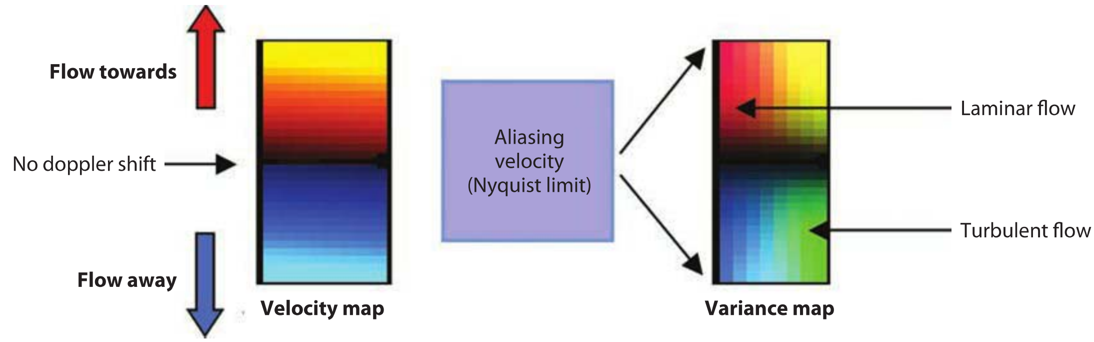

FIGURE 1–10. Characteristics of color flow maps.

of flow and the magnitude of the Doppler shift are displayed as color maps, which can be *velocity maps* or *variance maps* (Fig. 1-10). A variance map contains information on the quality of flow (i.e., laminar vs. turbulent); however, turbulent flow and signal aliasing will result in an apparent wide range of velocities. Also, in the case of color flow Doppler, aliasing may introduce confusion as to the direction of flow. Color flow and spectral Doppler are set as a high-pass filter to eliminate tissue motion artifacts.

A typical (but not uniform) convention for color Doppler velocity maps is for red to indicate flow toward the probe and for blue to indicate flow away from the probe (BART = Blue Away, Red Toward). A region that is black on color flow Doppler imaging represents an area where there is no measured Doppler shift.

BIOEFFECTS

US bioeffects include *thermal effects* and *cavitation*. In addition, mechanical effects (vibration) may be of concern. *Thermal bioeffects* consist of a temperature elevation resulting from the absorption and scattering of US by biologic tissue and is related to beam intensity (the spatial peak and temporal average [SPTA] intensity). The SPTA limits are 100 mW/cm^2 for unfocused beams and 1000 mW/cm^2 for focused beams. *Cavitation* results from the interaction of US with microscopic gas bubbles. Stable cavitation refers to forces that cause the bubbles to contract and expand. Transient cavitation results in breaking the bubbles and releasing energy, producing perhaps more pronounced effects on tissues at the microscopic level. The *mechanical index (MI)*, a calculated and unitless number, is used to convey the likelihood of bioeffects from cavitation. At low MI (<0.1) the microbubbles expand and contract in a linear fashion. High MI (>1) sound beams result in bubble disruption (extreme nonlinear behavior).

The U.S. Food and Drug Administration (FDA) limits the maximum intensity output of cardiac ultrasound systems to less than 720 W/cm^2 due to concerns of possible tissue and neurological damage from mechanical injury.

REVIEW QUESTIONS[1–3,5]

Basics of Ultrasound

Select the *one best* answer for each item.

1. Which of the following is *not* an acoustic variable?
   a. Pressure
   b. Density
   c. Distance
   d. Intensity

2. Which of the following sound wave frequencies is ultrasonic?
   a. 10 Hz
   b. 10 MHz
   c. 10 kHz
   d. 10,000 Hz

3. An increase in the strength of the US pulse will increase:
   a. Frequency
   b. Intensity
   c. Pulse duration
   d. Pulse repetition frequency

4. If imaging depth decreases, pulse repetition frequency:
   a. Decreases
   b. Does not change
   c. Increases
   d. Varies

5. An example of a Rayleigh scatterer is the:
   a. Red blood cell
   b. Kidney
   c. Mitral valve
   d. Pericardium

6. If the frequency is doubled, the period:
   a. Increases two-fold
   b. Decreases
   c. Does not change
   d. Increases ten-fold

7. The wavelength in soft tissue of sound with a frequency of 2 MHz is:
   a. 6.16 mm
   b. 3.08 mm
   c. 1.54 mm
   d. 0.77 mm

8. The speed of sound is slowest in:
   a. Air
   b. Fat
   c. Soft tissue
   d. Bone

9. Which of the following parameters of sound are determined by the sound source *and* the medium?
   a. Frequency
   b. Wavelength
   c. Amplitude
   d. Propagation speed

10. Reflection occurs when the two media at the boundary have:
    a. Identical acoustic impedances
    b. Different acoustic impedances
    c. Identical densities and propagation speeds
    d. Different temperatures

11. All of the following are true of refraction *except*:
    a. Is a change in direction of wave propagation when traveling from one medium to another
    b. Occurs when there are different propagation speeds and oblique incidence
    c. Is described by Snell's law
    d. Occurs with different propagation speeds and normal incidence

12. A sound beam strikes the boundary between two media at an incident angle of 45 degrees and is partly reflected and transmitted. If medium A has an impedance of 1.25 MRayls and a propagation speed of 1540 m/s and medium B has an impedance of 1.85 MRayls and a propagation speed of 2.54 km/s, what is the angle of *reflection?*
    a. 45 degrees
    b. 30 degrees
    c. 60 degrees
    d. 15 degrees

13. A sound beam strikes the boundary between two media at an incident angle of 45 degrees and is partly reflected and transmitted. If the propagation speed of the second medium is slower than the propagation speed of the first medium, then the transmission angle is:
    a. Equal to the incident angle
    b. Greater than the incident angle
    c. Less than the incident angle
    d. Cannot be determined

14. A sound wave leaves its source and travels through a liquid. If the speed of sound through that liquid is 600 m/s and the echo returns to the source 1 second later, at what distance is the source from the reflector?
    a. 1540 m
    b. 770 m
    c. 600 m
    d. 300 m

15. The amplitude of a wave is:
    a. The difference between the average and maximum (or minimum) values of an acoustic variable
    b. Determined initially by the medium
    c. Altered by the sonographer
    d. Twice the average amplitude

16. Intensity is inversely proportional to:
    a. Beam area
    b. Power
    c. Amplitude
    d. Amplitude squared

17. The speed of sound in a medium increases when:
    a. Elasticity of the medium increases
    b. Density of the medium increases
    c. Stiffness of the medium decreases
    d. Stiffness of the medium increases

18. Increasing the frequency of a transducer:
    a. Increases wavelength
    b. Improves axial resolution
    c. Increases depth of penetration
    d. Increases pulse duration

19. Propagation speed:
    a. Can be changed by the sonographer
    b. Is an average of 1540 km/s in soft tissue
    c. Is slower in a liquid than in a solid
    d. Is determined by the sound source

20. Attenuation of an ultrasound beam results from:
    a. Absorption
    b. Reflection