CHAPTER 7

MEDICAL IMAGING MODALITIES: ULTRASOUND IMAGING

Sound or acoustic waves were successfully used in sonar technology in military applications in World War II. The potential of ultrasound waves in medical imaging was explored and demonstrated by several researchers in the 1970s and 1980s, including Wild, Reid, Frey, Greenleaf, and Goldberg (1–7). Today ultrasound imaging is successfully used in diagnostic imaging of anatomical structures, blood flow measurements, and tissue characterization. Safety, portability, and low-cost aspects of ultrasound imaging have made it a successful diagnostic imaging modality (5–9).

7.1. PROPAGATION OF SOUND IN A MEDIUM

Sound waves propagate mechanical energy, causing periodic vibration of particles in a continuous elastic medium. Sound waves cannot propagate in a vacuum since there are no particles of matter in the vacuum. The initial energy creates the mechanical movement of a particle through compression and rarefaction that is propagated through the neighbor particles depending on the density and elasticity of the material in the medium.

Sound waves are characterized by wavelength and frequency. Sound waves audible to the human ear are comprised of frequencies ranging from 15 to 20 kHz. Sound waves with frequencies above 20 kHz are called ultrasound waves. The velocity of a sound wave c in a medium is related to its wavelength λ and frequency ν by

$$c = \lambda \nu. \tag{7.1}$$

Since the frequency remains constant in the medium even when there is a change in the medium such as from soft tissue to fat, the propagation speed or velocity of the sound determines the wavelength. For example, the velocity of a sound in air, water, soft tissue, and fat are, respectively, 331, 1430, 1540, and 1450 m/s. Thus, a 5 MHz ultrasound beam has a wavelength of 0.308 mm in soft tissue with a velocity

Medical Image Analysis, *Second Edition*, by Atam P. Dhawan of 1540 m/s. Because the frequency remains unchanged when a sound wave enters from one medium to another, the resultant change is experienced in the wavelength and the direction of the propagation determined by the principle of refraction. Like light waves, sound waves follow the principles of reflection, refraction, and superposition. The amplitude of a sound wave is the maximum particle displacement in the medium determined by the pressure applied by the sound energy. The intensity, I, is proportional to the pressure and is measured by power per unit area. For example, the intensity of sound waves can be expressed in milliwatts/per square centimeter (mW/cm^2). Another term, decibel (dB), is often used in expressing the relative intensity levels and is given by

$$\text{Relative Intensity in dB} = 10\log_{10}\frac{I_1}{I_2} \tag{7.2}$$

where I_1 and I_2 are intensity values of sound energy that are being compared.

For example, an incident ultrasound pulse of intensity I_1 loses its intensity when it travels through soft tissue. If its intensity in the soft tissue is I_2, the loss can be computed from Equation 7.2 and is expressed in decibels. Thus, a 100 mW/cm^2 acoustic pulse would be at 0.1 mW/cm^2 after a 30 dB loss.

As mentioned above, sound waves propagate through displacement of particles in a medium. Let us assume that the medium is linear and lossless to the propagation of a sound wave that applies a mechanical force ∂F to cause a particle displacement ∂u (in the range of a few nanometers) in the direction z (depth in the tissue while the x–y coordinates represent the two-dimensional [2-D] plane of the surface of the tissue). The particle displacement in the direction z can then be represented by a second-order differential equation as (1–5, 8)

$$\frac{\partial^2 u}{\partial z^2} = \frac{1}{c^2}\left(\frac{\partial^2 u}{\partial t^2}\right) \tag{7.3}$$

where c is the velocity of sound propagation in the medium that is assumed to be linear and lossless.

Equation 7.3 is known as the wave equation where the velocity of sound wave is given by

$$c = \sqrt{\frac{1}{k\rho}} \tag{7.4}$$

where k and ρ are, respectively, the compressibility (inverse of the bulk modulus) and density of the medium.

The solution to the second-order differential wave equation (Eqn 7.3) can be given by an appropriate selection of the function $u(t, z)$ as a function of time t and space z as

$$u(t,z) = u_0 e^{j\omega(ct-z)} \text{ with } \omega = \frac{2\pi}{\lambda} \tag{7.5}$$

where ω and λ represent, respectively, wave number and wavelength.

The particle displacement causes a pressure wave $p(t, z)$ in the medium that can be represented by

$$p(t,z) = p_0 e^{j\omega(ct-z)}. \tag{7.6}$$

It can be realized that

$$u = \frac{p}{Z} \tag{7.7}$$

where Z represents the acoustic impedance of the medium at the point of propagation.

The pressure at a particular point is related to the displacement and density of the medium by

$$p = \rho cu. \tag{7.8}$$

From Equations 7.7 and 7.8, it follows that

$$Z = \rho c. \tag{7.9}$$

Thus, the acoustic impedance of a medium is defined as the product of the density and sound propagation velocity in the medium. Since the intensity is also defined by the average flow of energy per unit area in the medium perpendicular to the direction of propagation, the intensity of the propagating wave is given as

$$I = \frac{p_0^2}{2Z}. \tag{7.10}$$

The density and elastic properties of the medium play a significant role in the propagation of sound. Stress–strain relationships of the medium are used to define elasticity and compressibility and can be described in a mathematical form to show the precise dependency of the propagation of sound waves in an inhomogeneous medium. However, the common properties of the propagation of sound waves in terms of reflection, refraction, scattering, and attenuation are similar to those experienced with light waves. Ultrasound waves can be reflected, refracted, and attenuated depending on the changes in the acoustic impedance in the medium.

In medical imaging applications, shorter wavelengths provide deeper penetration and better spatial resolution. Since the velocity of sound in a specific medium is fixed, the wavelength is inversely proportional to the frequency. In medical ultrasound imaging, sound waves of 2–10 MHz can be used, but 3–5 MHz frequencies are the most common. Higher frequencies are used for low-depth high-resolution imaging.

7.2. REFLECTION AND REFRACTION

As the incident sound waves reach a boundary surface that defines a change of the medium with an angle not perpendicular to the surface, a part of the acoustic energy is reflected and the remaining is transmitted into the second medium. With the change in medium, the transmitted wave changes its direction. Like the refraction

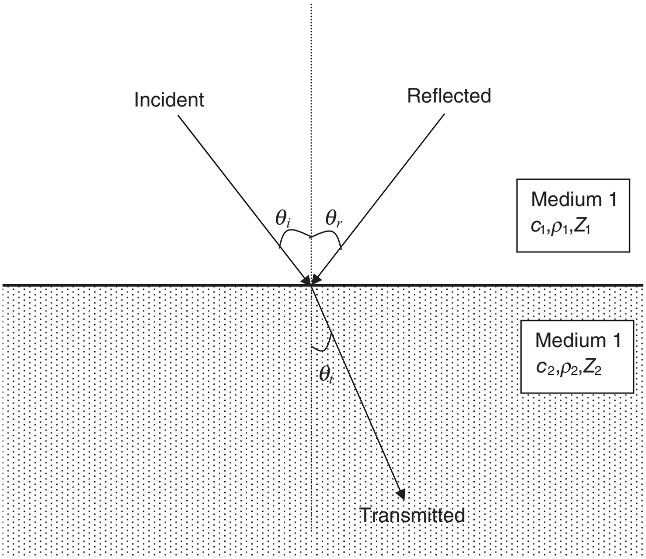

Figure 7.1 Reflection and refraction of ultrasound waves as incident from medium 1 to medium 2.

phenomenon of light waves, the angle of incidence is equal to the angle of reflection (Fig. 7.1). The angle of a transmitted wave is given by Snell's law and is dependent on the propagation speeds of the two mediums. Let us assume that c_1 and c_2 are, respectively, the propagation speeds of the incident sound wave in the two mediums across the incident boundary. According to Snell's law,

$$\theta_i = \theta_r$$
$$\frac{c_1}{c_2} = \frac{\sin\theta_i}{\sin\theta_t} \tag{7.11}$$

where θ_i and θ_t are the incident and transmitted waves, respectively, in medium 1 and medium 2 (Fig. 7.1).

It should be noted that a critical incident angle θ_{ci} exists for which the incident wave is completely reflected without any transmission as

$$\theta_{ci} = \sin^{-1}\left(\frac{c_1}{c_2}\right) \text{ with } c_2 > c_1. \tag{7.12}$$

7.3. TRANSMISSION OF ULTRASOUND WAVES IN A MULTILAYERED MEDIUM

When a sound wave enters from one medium to another such as in the case of a boundary of a tissue, the acoustic impedance changes and therefore the direction of the wave is altered. The difference in the acoustic impedance of the two media causes reflection of a fraction of the incident sound energy. If a sound wave in incident with an angle θ_i at the interface of two mediums with acoustic impedance, Z_1 and Z_2, the pressure reflection coefficient or simply called reflection coefficient R is given by (1–3, 8)

$$R = \frac{p_r}{p_i} = \frac{Z_2 \cos\theta_i - Z_1 \cos\theta_t}{Z_2 \cos\theta_i + Z_1 \cos\theta_t}. \tag{7.13}$$

As the pressure waves generated by reflection and transmission are in opposite direction, it can be realized that

$$p_t = p_i + p_r. \tag{7.14}$$

Thus, the transmitted pressure coefficient T is given by

$$T = \frac{p_t}{p_i} = \frac{p_i + p_r}{p_i} = 1 + \frac{p_r}{p_i} = 1 + R$$

which gives

$$\mathrm{T} = \frac{2Z_2 \cos\theta_i}{Z_2 \cos\theta_i + Z_1 \cos\theta_t}. \tag{7.15}$$

Using the above formulations, intensity reflection coefficients R_I and intensity transmission coefficients T_I can be expressed as

$$\begin{aligned} R_I &= \frac{I_r}{I_i} = R_p^2 = \frac{(Z_2 \cos\theta_i - Z_1 \cos\theta_t)^2}{(Z_2 \cos\theta_i + Z_1 \cos\theta_t)^2} \\ T_I &= 1 - R_I = \frac{4Z_2 Z_1 \cos^2\theta_i}{Z_2 \cos\theta_i + Z_1 \cos\theta_t}. \end{aligned} \tag{7.16}$$

If the incident ultrasound wave is focused at the interface of medium 1 and medium 2 in the direction perpendicular to the boundary, the above expressions for reflection and transmission coefficients can be simplified as

$$R_{1,2} = \frac{Z_2 - Z_1}{Z_1 + Z_2}$$

and

$$T_{1,2} = \frac{2Z_2}{Z_1 + Z_2} \tag{7.17}$$

where $R_{1,2}$ is the reflection from the second medium to the first medium and $T_{1,2}$ is the transmission from the first medium to the second medium.

In medical ultrasound imaging, the sound waves propagate through multilayered structures. A simplified multilayered structure with acoustic impedance Z_i; $i = 1, 5$ is shown in Figure 7.2. Generalizing the above equations for normal incidence cases, the transmission and reflection coefficients can be expressed as

$$R_{ij} = \frac{Z_j - Z_i}{Z_i + Z_j}$$

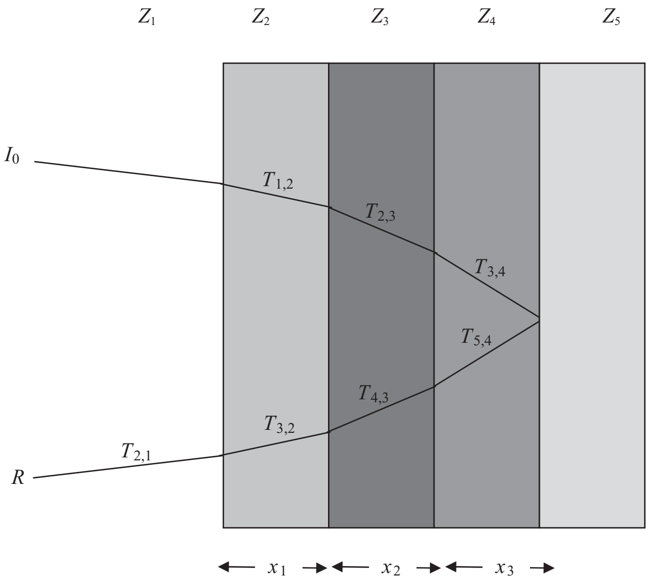

Figure 7.2 A path of a reflected sound wave in a multilayered structure.

and

$$T_{ij} = \frac{2Z_j}{Z_i + Z_j}. \tag{7.18}$$

Applying Equation 7.18 to the multilayer structure shown in Figure 7.2, the final reflected wave can be expressed as

$$R_0 = I_0 T_{12} T_{23} T_{34} T_{54} T_{43} T_{32} T_{21}. \tag{7.19}$$

Since $1 + R_{ij} = T_{ij}$, the above can be simplied as

$$R_0 = I_0(1 - R_{12}^2)(1 - R_{23}^2)(1 - R_{34}^2)R_{45}. \tag{7.20}$$

7.4. ATTENUATION

Various mechanical properties such as density, elasticity, viscosity, and scattering properties cause attenuation of the sound waves in a medium. As with electromagnetic radiation, the loss of energy of sound waves occurs as the waves propagate in a medium. Extending Equation 7.5 to include viscosity and compressibility of the medium, the displacement can be related to the attenuation coefficient α of the ultrasound wave as (1, 8, 9)

$$u(t, z) = u_0 e^{-\alpha z} e^{j\omega(ct - z)}$$
$$\text{with } \alpha = \frac{\left(\frac{4\gamma}{3} + \xi\right)\omega^2}{2\rho c} \tag{7.21}$$

TABLE 7.1 Attenuation Coefficients and Propagation Speeds of Sound Waves

Tissue	Average attenuation coefficient in dB/cm at 1 MHz	Propagation velocity of sound in m/s	Average acoustic impedance in $\times 10^5$ (g/cm^2/s)
Fat	0.6	1450	1.38
Soft tissue	0.7–1.7	1540	1.7
Liver	0.8	1549	1.65
Kidney	0.95	1561	1.62
Brain	0.85	1541	1.58
Blood	0.18	1570	1.61
Skull and bone	3–10 and higher	3500–4080	7.8
Air	10	331	0.0004

where γ and ξ, respectively, are shear viscosity and compressional viscosity coefficients.

As defined in Equation 7.10, the intensity of the propagating wave can now be expressed in terms of attenuation coefficient α as

$$I(t,z) = \frac{p_0{}^2}{Z} e^{-2\alpha z} e^{2j\omega(ct-z)}. \tag{7.22}$$

The attenuation coefficient is characterized in units of decibels per centimeter and is dependent on the frequency of the sound waves. As frequency increases, the attenuation coefficient also increases. Table 7.1 shows attenuation coefficients of some biological tissues.

7.5. ULTRASOUND REFLECTION IMAGING

Let us assume that a transducer provides an accoustic signal of $s(x, y)$ intensity with a pulse $\phi(t)$ that is transmitted in a medium with an attenuation coefficient, μ, and reflected by a biological tissue of reflectvity $R(x, y, z)$ with a distance z from the transducer. The recorded reflected intensity of a time varying accoustic signal, $J_r(t)$ over the region $\Re$ can then be expressed as

$$J_r(t) = K \left| \iiint_{\Re} \left(\frac{e^{-2\mu z}}{z} \right) R(x,y,z) s(x,y) \phi\left(t - \frac{2z}{c} \right) dx\, dy dz \right| \tag{7.23}$$

where K, $\phi(t)$, and c, respectively, represent a normalizing constant, received pulse, and the velocity of the acoustic signal in the medium.

Using an adaptive time-varying gain to compensate for the attenuation of the signal, Equation 7.23 for the compensated recorded reflected signal from the tissue, $J_{cr}(t)$, can be simplified to

$$J_{cr}(t) = K \left| \iiint_{\Re} R(x,y,z) s(x,y) \phi\left(t - \frac{2z}{c} \right) dx\, dy\, dz \right|$$

or, in terms of a convolution, as

$$J_{cr}(t) = K\left|R\left(x, y, \frac{ct}{2}\right) \otimes s(-x, -y)\phi(t)\right| \tag{7.24}$$

where ⊗ represents a three-dimensional (3-D) convolution. This is a convolution of a reflectivity term characterizing the tissue and an impulse response characterizing the source parameters.

7.6. ULTRASOUND IMAGING INSTRUMENTATION

A piezoelectric crystal-based transducer can be used as a source to form an ultrasound beam as well as a detector to receive the returned signal from the tissue. A schematic diagram of a typical single crystal-based ultrasound transducer is shown in Figure 7.3. In a plastic casing, a piezoelectric crystal is used along with a damping material layer and acoustic insulation layer inside the plastic casing. An electromagnetic tuning coil is used to apply a controlled voltage pulse to produce ultrasound waves. In the receiver mode, the pressure wave of the returning ultrasound signal is used to create an electric signal through the tuned electromagnetic coil. The piezoelectric crystal has a natural resonant frequency f_0 that can be expressed as

$$f_0 = \frac{c_1}{2d} \tag{7.25}$$

where c_1 is the velocity of sound in the crystal material and d is the thickness of the crystal.

While higher harmonics of the natural resonant frequency of the piezoelectric crystal can be created, multiple crystals can be grouped together to form a specific

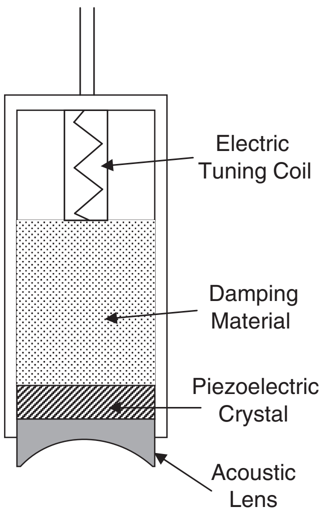

Figure 7.3 A schematic diagram of an ultrasound single-crystal transducer.

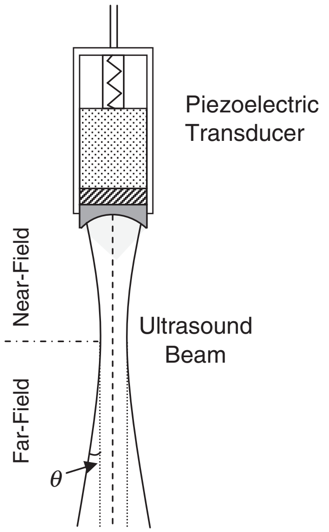

Figure 7.4 Ultrasound beam from a piezoelectric transducer.

ultrasound beam of desired shape and focus. The ultrasound beam (as shown in Fig. 7.4) transmits energy into three zones: (1) a near-field transmission (also called Fresnel region) of the energy of the ultrasound pulse; (2) a far-field transmission (also called Fraunhofer region); and (3) an unwanted side-lobe resulting from the radial expansion of the piezoelectric crystal appearing on the sides of the main direction of the beam. The length of the near-field region L_{nf} is determined by the radius of the crystal r and the wavelength λ of ultrasound beam as

$$L_{nf} = \frac{r^2}{\lambda}. \tag{7.26}$$

The far-field effect of the ultrasound beam is diverging as shown in Figure 7.4 with an angle that is given by

$$\theta = \arcsin\left(\frac{0.61\lambda}{r}\right). \tag{7.27}$$

A conventional ultrasound imaging system is comprised of a piezoelectric crystal-based transducer that works as a transmitter as well as receiver based on an electronic transmitter/receiver switching circuit, a control panel with pulse generation and control, and a computer processing and display system. Figure 7.5 shows a schematic diagram of a conventional ultrasound imaging system.

In the transmitter mode, a piezoelectric crystal converts an electrical signal to sound energy. The same crystal can convert sound waves into an electrical signal in the receiver mode. The transducers for medical imaging are tuned for the specific frequency range (2–10 MHz). A backing block absorbs the backward-directed vibrations and sound energy to eliminate undesired echoes from the housing enclosure of the piezoelectric crystal. Additional acoustic absorbers are used to improve the performance of ultrasound pulses with the desired mode and timings.

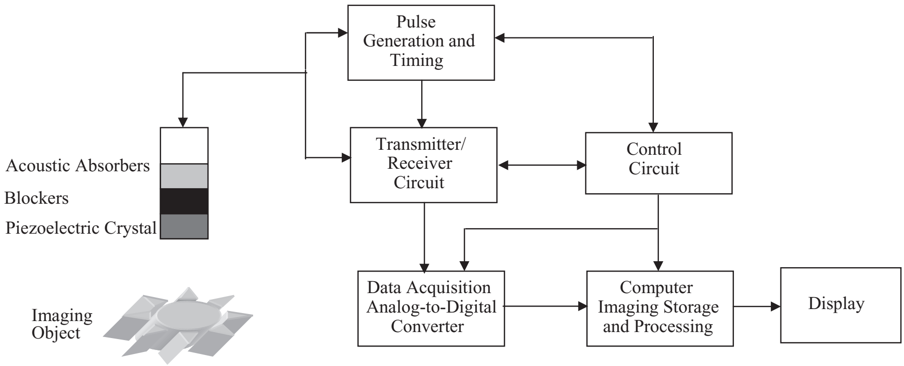

Figure 7.5 A schematic diagram of a conventional ultrasound imaging system.

Usually, an ultrasound transducer provides brief pulses of ultrasound when stimulated by a train of voltage spikes of 1–2 μs duration applied to the electrodes of the crystal element. An ultrasound pulse can be a few cycles long, typically on the order of two to three cycles. As the same crystal element is used as the receiver, the time between two pulses is used for detecting the reflected signal or echo from the tissue. The PRF is thus dependent on the time to be devoted to listening for the echo from the tissue. In a typical mode of operation, the transducer is positioned at the surface of the body or an object. The total travel distance traveled by the ultrasound pulse at the time of return to the transducer is twice the depth of the tissue boundary from the transducer. Thus, the maximum range of the echo formation can be determined by the speed of sound in the tissue multiplied by half of the pulse-repetition period. When the echoes are received by the transducer crystal, their intensity is converted into a voltage signal that generates the raw data for imaging. The voltage signal then can be digitized and processed according to the need to display on a computer monitor as an image.

Since the acoustic echoes are position-dependent, the spatial resolution and quality of ultrasound images are somewhat different from the conventional radiographic images. Typically, ultrasound images appear noisy with speckles, lacking a continuous boundary definition of the object structure. The interpretation and quantification of the object structure in ultrasound images is more challenging than in X-ray computed tomography (X-ray CT) or magnetic resonance (MR) images.

7.7. IMAGING WITH ULTRASOUND: A-MODE

The "A-mode" of ultrasound imaging records the amplitude of returning echoes from the tissue boundaries with respect to time. In this mode of imaging, the ultrasound pulses are sent in the imaging medium with a perpendicular incident angle. Unlike X-ray CT and MR imaging (MRI), the operator in ultrasound imaging has a great ability to control the imaging parameters in real time. These parameters include positioning, preamplification, time gain compensation and rejection of noisy echoes to improve the signal-to-noise ratio (SNR), leading to an improvement in image quality.

A-mode-based data acquisition is the basic method in all modes of diagnostic ultrasound imaging. Depending on the objectives of imaging and tissue characterization and the nature of the imaging medium, only relevant information from the A-mode-based signal acquisition is retained for further processing and display.

Since the echo time represents the acoustic impedance of the medium and depth of the reflecting boundary of the tissue, distance measurements for the tissue structure and interfaces along the ultrasound beam can be computed. This is a great advantage in ultrasound imaging. The intensity and time measurements of echoes can provide useful 3-D tissue characterization. Thus, shape and position measurements can be obtained from ultrasound images.

7.8. IMAGING WITH ULTRASOUND: M-MODE

The "M-mode" of ultrasound imaging provides information about the variations in signal amplitude due to object motion. A fixed position of the transducer, in a sweep cycle, provides a line of data that is acquired through A-mode. The data is displayed as a series of dots or pixels with brightness level representing the intensity of the echoes. In a series of sweep cycles, each sequential A-line data is positioned horizontally. As the object moves, the changes in the brightness levels representing the deflection of corresponding pixels in the subsequent sequential lines indicate the movement of the tissue boundaries. Thus, the x-axis represents the time, while the y-axis indicates the distance of the echo from the transducer.

M-mode is a very effective method in tracking the object motion such as the movement of a valve in a beating heart. Figure 7.6 shows the M-mode display of an ultrasound cardiac signal showing mitral stenosis with anterior motion of the posterior mitral leaflet.

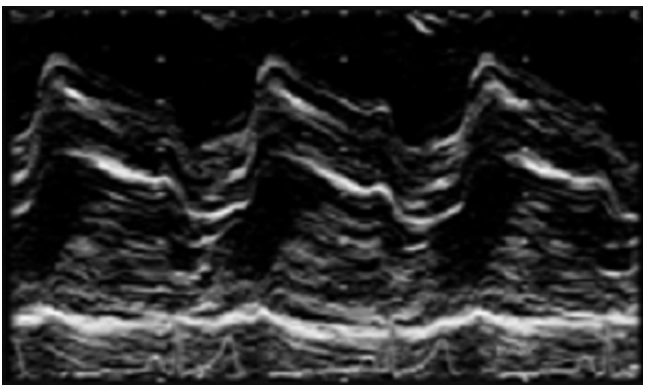

Figure 7.6 M-Mode display of mitral valve leaflet of a beating heart.

7.9. IMAGING WITH ULTRASOUND: B-MODE

The “B-Mode” of ultrasound imaging provides 2-D images representing the changes in acoustic impedance of the tissue. The brightness of the B-Mode image shows the strength of the echo from the tissue structure. To obtain a 2-D image of the tissue structure, the transducer is pivoted at a point about an axis and is used to obtain a V-shaped imaging region. Alternately, the transducer can be moved to scan the imaging region. Using position encoders and computer processing, several versions of the acquired data can be displayed to show the acoustic characteristics of the tissue structure and its medium.

As with the A-mode method, the returned echoes are processed with proper preamplification and adaptive gain amplifiers for acquiring the raw data that are converted into a 2-D image for display. Dynamic B-mode scanners provide real-time ultrasound images using multiple transducer arrays and computer control-based data acquisition and display systems. Several types of transducer arrays such as linear, annular, and linear phased arrays can be used in dynamic real-time ultrasound scanners. In such array-based scanners, the ultrasound beam is electronically steered with pulse generators that are synchronized with the clock for automatic data acquisition. The operator-based dependency is thus minimized in such ultrasound imaging systems. Figure 7.7 shows a B-scan of the mitral valve area of a beating heart.

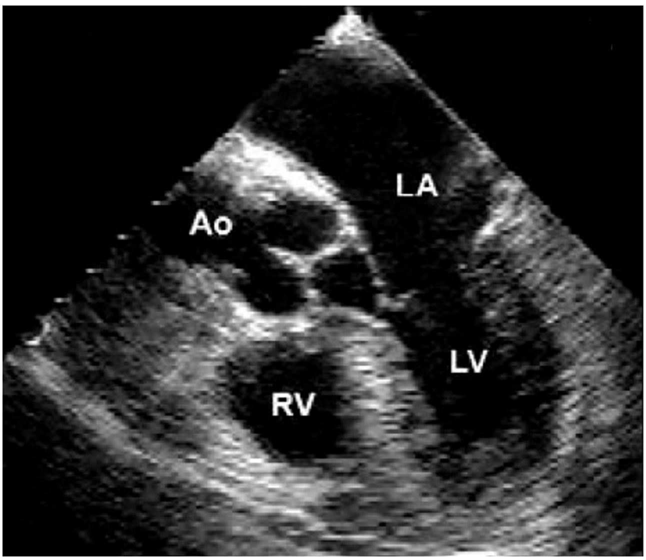

Figure 7.7 The B-Mode image of a beating heart for investigation of mitral stenosis. Left atrium (LA), left ventricle (LV), right ventricle (RV), and aorta (Ao) are labeled in the image.

7.10. DOPPLER ULTRASOUND IMAGING

Blood flow can be effectively imaged with ultrasound imaging using the Doppler method. According to the Doppler effect, a change in the frequency from a moving source is observed by a stationary observer. The change in the perceived or observed frequency is called the Doppler frequency, $f_{doppler}$, and is given by

$$f_{doppler} = \pm \frac{2v \ \cos\theta \ f}{c+v} \tag{7.28}$$

where v is the velocity of the moving source or object, f is the original frequency, c is the velocity of the sound in the medium, and θ is the incident angle of the moving object with respect to the propagation of the sound.

Since the velocity of the moving object is much less than the velocity of the sound, the above equation is simplified in medical imaging as

$$f_{doppler} = \pm \frac{2v \ \cos\theta \ f}{c}. \tag{7.29}$$

Since the above term of Doppler shift includes a cosine of the angle of incidence wave, it should be noted that there will be no shift in Doppler frequency if the angle of incidence is 90 degrees. The Doppler shift is a very popular method of measuring blood flow with ultrasound pulses typically at 5 MHz. It is not possible to align the ultrasound transducer parallel to the flow at 90-degree incidence angle. The preferred angles are between 30 and 60 degrees to obtain a good discriminatory signal for Doppler shift.

In Doppler scanners, the received signal is amplified and mixed with the reference frequency (transmitter signal). The demodulated signal is then passed through a series of filters zero-crossing detectors to estimate the deviation from the reference signal. Thus, the Doppler shift representing the flow is measured and displayed against the time. With a single transducer, a pulse-echo format is used for Doppler ultrasound imaging. A spatial pulse width of usually 5–25 cycles is used to achieve a narrow-band frequency response. Depth selection is achieved through electronic gating of the echo time. For a given depth, the round-trip echo time is determined and used to select the correct echo from the selected depth, as all other echoes are rejected in the gating process. The sampling of the echo signal with the selection of a specific depth is accomplished by the operator who manipulates the pulse-repetition frequency (PRF) to be greater than the twice the maximum Doppler frequency shift (following the Nyquist criterion) for accurate measurements. Alternately, two transducers can be used in a continuous mode where one transducer transmits the incident ultrasound waves and the second continuously receives the echoes. Using the demodulator and spectrum analyzer, all similar frequencies from stationary objects are filtered out to extract the Doppler shift information from the returned echoes.

Modern imaging systems convert the Doppler shift into a 2-D image through spatial scanning and mapping of the Doppler frequency shift into color or brightness levels. Figure 7.8 shows a diastolic color Doppler flow convergence in the apical four-chamber view of mitral stenosis.

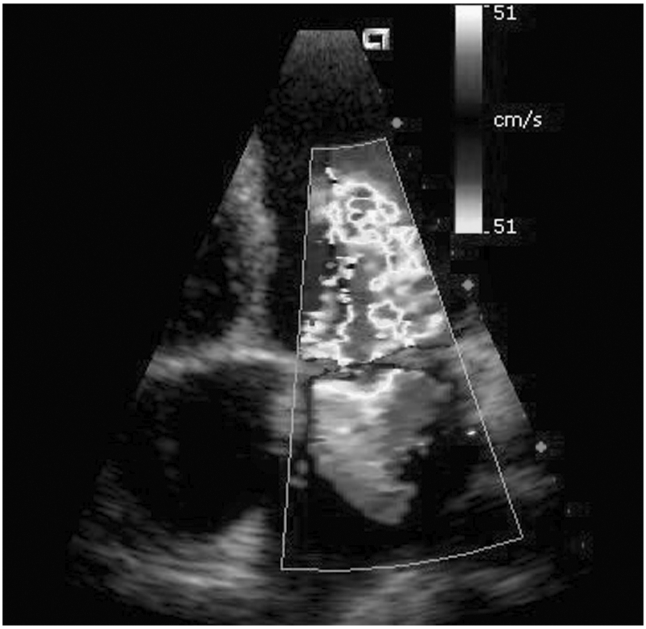

Figure 7.8 A Doppler image of the mitral valve area of a beating heart.

7.11. CONTRAST, SPATIAL RESOLUTION, AND SNR

Contrast in ultrasound images depends on the ability of detection of small differences in acoustic signals for discrimination between the tissue types. Since ultrasound imaging is based on the returned echoes, the contrast is determined by the SNR in detection of weak echoes. Contrast can be improved by increasing the intensity or power of the ultrasound source and also by increasing the PRF, as it allows better lateral resolution and signal averaging. However, any ultrasound imaging system may have limitations on increasing the power level as well as the PRF depending on the piezoelectric crystal, amplifier and mixer/demodulator subsystems, and depth of scanning. Moreover, interferences of multiple echoes from other objects in the turbid biological medium cause considerable noise and artifacts in the received signal (returned echoes).

The spatial resolution in ultrasound images is analyzed in axial and lateral directions. The axial resolution (also called range, longitudinal, or depth resolution) is determined by the spatial discrimination of two separate objects in the direction parallel to the incident ultrasound beam. The depth positions of reflecting surfaces or objects have to be resolved through returned echoes. The minimum separation distance between reflecting surfaces is determined by the half of the spatial pulse width, which is represented by the Q factor. Thus, a narrower pulse width should be used to improve axial resolution. For example, an ultrasound beam at 5 MHz may provide an axial resolution up to 0.3 mm with appropriate electronic subsystems.

A lateral resolution (also called azimuthal resolution) is determined by the ability to discriminate two spatial objects perpendicular to the direction of the incident ultrasound beam. Lateral resolution depends on the diameter of the ultrasound beam. As shown in Figure 7.4, the diameter of an ultrasound beam varies with the distance between the transducer and object. In the near-field (Fresnel region), the best lateral resolution can be considered equal to the radius of the transducer. As the beam enters into the far-field or Fraunhofer region, the lateral resolution worsens with the diverging diameter of the beam. To improve lateral resolution, acoustic lenses can be used to focus the beam and reduce its diameter. It can be noted that in order to reduce the diameter of the beam, the range or focus of the beam has to be sacrificed for a smaller depth. Phased array transducers can be used to achieve variable depths of focus in the lateral direction (5, 7).

In dynamic imaging mode, the PRF is determined by the product of frame rate and number of scan lines per image. Increasing the PRF provides an increased number of scan lines per image for better lateral resolution. However, as mentioned above, the increase in PRF is limited because of the depth of scanning and the frequency response of the associated hardware subsystems.

The SNR of an ultrasound imaging system depends on the differentiation of the meaningful echoes from the noise that may be created by the interferences of multiple echoes from the surrounding objects and also by the system components. The first type of noise, because of interferences of coherent waves in the medium, appears as speckles in the image. The electronics associated with the detection system adds noise to ultrasound images, causing appearances of noisy speckles with granular structures. In addition, the spectral response of the ultrasound transducer and the side lobes also contribute to the noise and clutter in the signal. Increasing the frequency of the ultrasound source increases the attenuation coefficients and lowers the SNR for higher depths of penetration. An acoustic focusing lens improves the SNR within the near-field region, as discussed above, but lowers the SNR outside the focus regions because of sharp divergence in the far-field (Franhaufer region). As discussed above, to improve the SNR of an ultrasound imaging system, the intensity and PRF should be increased. However, an increase in power and PRF may be limited due to the depth of scanning and limitations on the associated hardware.

7.12. EXERCISES

7.1. Describe the fundamental principles of ultrasound imaging that make it capable of tissue characterization.

7.2. For ultrasound imaging at 3 MHz, what will be the wavelength in a soft tissue? Assume that the speed of ultrasound in soft tissue is 1540 m/s.

7.3. Describe what changes will be experienced when an ultrasound wave of 5 MHz will travel through fat and then enter into kidney. Assume the speed of ultrasound in fat and kidney, respectively, is 1450 m/s and 1561 m/s.

7.4. Describe the desired characteristics of an ultrasound transducer.

7.5. If an ultrasound crystal is 1.8 mm thick, considering that the speed of sound in the crystal is 4000 m/s, what is the natural frequency generated by the ultrasound transducer?

7.6. What are the significant factors in determining the contrast of ultrasound images?

7.7. What factors can improve signal-to-noise ratio (SNR) in ultrasound B-mode imaging?

7.8. Describe the M-mode operation for ultrasound imaging.

7.9. What is the Doppler effect and how is it used in ultrasound imaging?

7.10. Consider an ultrasound imaging system at 5 MHz that is used in imaging a blood flow with a velocity of 30 cm/s. If the ultrasound transducer is targeted at blood vessels at 45 degrees, calculate the Doppler shift.

7.11. Can an acoustic lens provide a higher SNR in ultrasound imaging for near-field and far-field regions? Explain your answer.

7.12. What is the limit of axial resolution in the near-field region for a 4 MHz ultrasound beam?

7.13. Display an ultrasound image of a cardiac cavity in MATLAB. Also display an X-ray CT image of the cardiac region. Apply an edge enhancement operation on both images. Can you define structural boundaries in ultrasound image? Explain your answer.

7.14. Display an ultrasound image of a cardiac cavity in MATLAB. Threshold the image using various gray levels of 50, 75, 100, 150, 200, and 225. Do you see any meaning meaningful region from thresholded images? Explain your answer.

7.13. REFERENCES

1. J.T. Bushberg, J.A. Seibert, E.M. Leidholdt, and J.M. Boone, *The Essentials of Medical Imaging*, Williams & Wilkins, Philadelphia, 1994.
2. Z.H. Cho, J.P. Jones, and M. Singh, *Fundamentals of Medical Imaging*, John Wiley & Sons, New York, 1993.
3. P. Fish, *Physics and Instrumentation of Diagnostic Medical Ultrasound*, John Wiley & Sons, Chichester, 1990.
4. P.N.T. Wells, *Biomedical Ultrasonics*, Medical Physics Series, Academic Press, London, 1977.
5. F.W. Kremkau, *Diagnostic Ultrasound Principles and Instrumentation*, Saunders, Philadelphia, 1995.
6. F.W. Kremkau, *Doppler Ultrasound: Principles and Instruments*, Saunders, Philadelphia, 1991.
7. D. Hykes, *Ultrasound Physics and Instrumentation*, Mosby, New York, 1994.
8. E. Konofagou, "Principles of ultrasound imaging," in A.P. Dhawan, H.K. Hunag, and D.S. Kim (Eds), *Principal and Advanced Methods in Medical Imaging and Image Analysis*, World Scientific Press, Singapore, pp. 129–149, 2008.
9. A. Webb, *Introduction to Biomedical Imaging*, IEEE-Wiley Interscience, Hoboken, 2003.