

Pat F. Fulgham

Introduction

The use of ultrasound is fundamental to the practice of urology. In order for urologists to best use this technology on behalf of their patients, they must have a thorough understanding of the physical principles of ultrasound. Understanding how to tune the equipment and to manipulate the transducer to achieve the best-quality image is crucial to the effective use of ultrasound. The technical skills required to perform and interpret urologic ultrasound represent a combination of practical scanning ability and knowledge of the underlying disease processes of the organs being imaged. Urologists must understand how ultrasound affects biological tissues in order to use this modality safely and appropriately. When the physical principles of ultrasound are fully understood, urologists will recognize both the advantages and limitations of ultrasound.

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The Mechanics of Ultrasound Waves

The image produced by ultrasound is the result of the interaction of mechanical ultrasound waves with biologic tissues and materials. Because ultrasound waves are transmitted at frequent intervals and the reflected waves received by the transducer, the images can be reconstructed and refreshed rapidly, providing a real-time image of the organs being evaluated. Ultrasound waves are **mechanical waves** which require a physical medium (such as tissue or fluid) to be transmitted. Medical ultrasound imaging utilizes frequencies in the one million cycles per second (or MHz) range. Most transducers used in urology vary from 2.5 to 18 MHz, depending on the application.

Ultrasound waves are created by applying alternating current to piezoelectric crystals within the transducer. Alternating expansion and contraction of the piezoelectric crystals creates a mechanical wave which is transmitted through a coupling medium (usually gel) to the skin and then into the body. The waves that are produced are **longitudinal waves**. This means that the particle motion is in the same direction as the propagation of the wave (Fig. 2.1). This longitudinal wave produces areas of rarefaction and compression of tissue in the direction of travel of the ultrasound wave.

The compression and rarefaction of molecules is represented graphically as a sine wave alternating between a positive and negative

Fig. 2.1 Longitudinal waves. The expansion and contraction of piezoelectric crystals caused by the application of alternating current to the crystals causes compression and rarefaction of molecules in the body

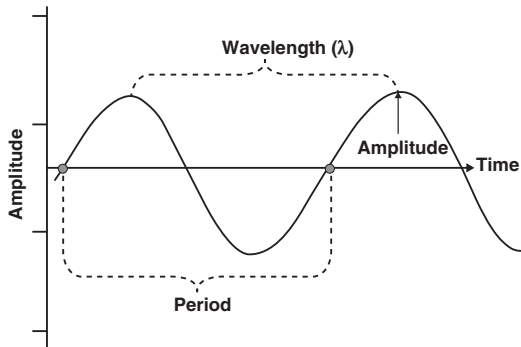
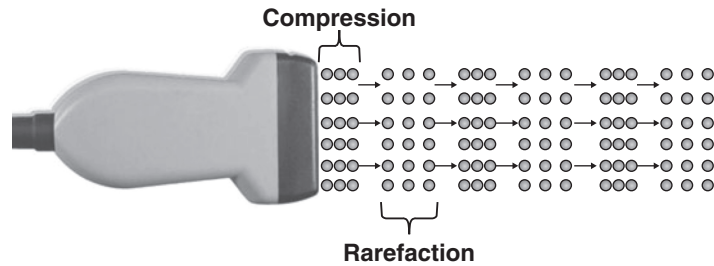


Fig. 2.2 Characteristics of a sound wave: the amplitude of the wave is a function of the acoustical power used to generate the mechanical compression wave and the medium through which it is transmitted

deflection from the baseline. A **wavelength** is described as the distance between one peak of the wave and the next peak. One complete path traveled by the wave is called a **cycle**. One cycle per second is known as 1 Hz (Hertz). The **amplitude** of a wave is the maximal excursion in the positive or negative direction from the baseline, and the **period** is the time it takes for one complete cycle of the wave (Fig. 2.2).

The velocity with which a sound wave travels through tissue is a product of its frequency and its wavelength. The velocity of sound in tissues is constant. Therefore, as the frequency of the sound wave changes, the wavelength must also change. The average velocity of sound in human tissues is 1540 m/s. Wavelength and frequency vary in an inverse relationship. Velocity equals frequency times wavelength (Fig. 2.3). As the frequency diminishes from 10 to 1 MHz, the wavelength increases from 0.15 to 1.5 mm. This has important consequences for the choice of transducer depending on the indication for imaging.

$$v = f\lambda$$

velocity = frequency x wavelength

Fig. 2.3 Since the velocity of sound in tissue is a constant, the frequency and wavelength of sound must vary inversely

Ultrasound Image Generation

The image produced by an ultrasound machine begins with the transducer. **Transducer** comes from the Latin **transducere**, which means to convert. In this case, electrical impulses are converted to mechanical sound waves via the **piezoelectric effect**.

In ultrasound imaging the transducer has a dual function as a sender and a receiver. Sound waves are transmitted into the body where they are at least partially reflected. The piezoelectric effect occurs when alternating current is applied to a crystal containing dipoles [1]. Areas of charge within a piezoelectric element are distributed in patterns which yield a “net” positive and negative orientation. When alternating charge is applied to the two element faces, a relative contraction or elongation of the charged areas occurs resulting in a mechanical expansion and then a contraction of the element. This results in a mechanical wave which is transmitted to the patient (Fig. 2.4).

Reflected mechanical sound waves are received by the transducer and converted back into electrical energy via the piezoelectric effect. The electrical energy is interpreted within the ultrasound instrument to generate an image which is displayed upon the screen.

For most modes of ultrasound, the transducer emits a limited number of wave cycles (usually two to four) called a pulse. The frequency of the two to

Fig. 2.4 Piezoelectric effect. Areas of “net” charge within a crystal expand or contract when current is applied to the surface, creating a mechanical wave. When the returning wave strikes the crystal, an electrical current is generated

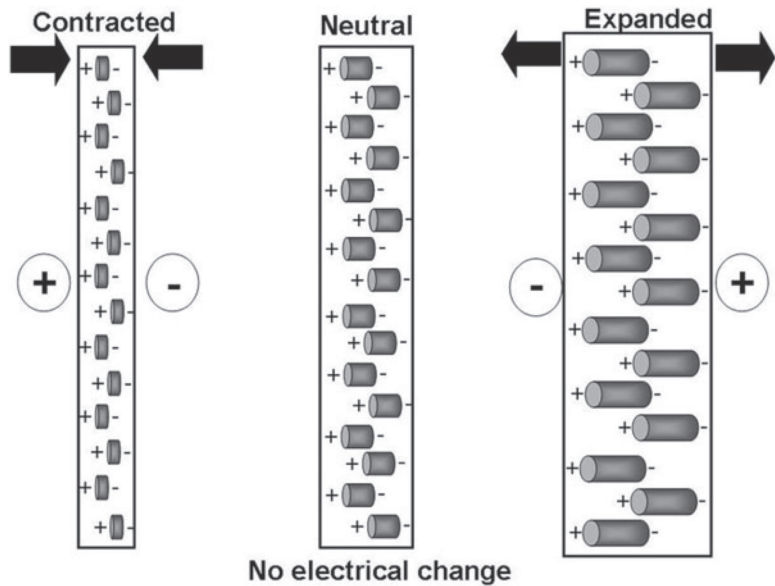
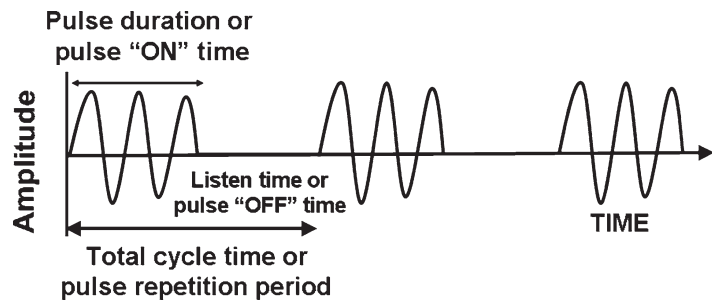


Fig. 2.5 The pulsed-wave ultrasound mode depends on an emitted pulse of 2–4 wave cycles followed by a period of “silence” as the transducer awaits the return of the emitted pulse



four waves within each cycle is usually in the 2.5–14 MHz range. The transducer is then “silent” as it awaits the return of the reflected waves from within the body (Fig. 2.5). The transducer serves as a receiver more than 99 % of the time.

Pulses are sent out at regular intervals usually between 1 and 10 kHz which is known as the **pulse repetition frequency (PRF)**. By timing the pulse from transmission to reception, it is possible to calculate the distance from the transducer to the object reflecting the wave. This is known as **ultrasound ranging** (Fig. 2.6). This sequence is known as **pulsed-wave ultrasound**.

The amplitude of the returning waves determines the brightness of the pixel assigned to the reflector in an ultrasound image. The greater the amplitude of the returning wave, the brighter the pixel assigned. Thus, an ultrasound unit produces an “image” by first causing a transducer to emit a series of ultrasound waves at specific frequencies and intervals

and then interpreting the returning echoes for duration of transit and amplitude. This “image” is rapidly refreshed on a monitor to give the impression of continuous motion. Frame refresh rates are typically 12–30 per second. The sequence of events depicted in Fig. 2.7 is the basis for all “scanned” modes of ultrasound including the familiar gray-scale ultrasound.

Interaction of Ultrasound with Biological Tissue

As ultrasound waves are transmitted through human tissue, they are altered in a variety of ways including loss of energy, change of direction, and change of frequency. An understanding of these interactions is necessary to maximize image quality and correctly interpret the resultant images.

Fig. 2.6 Ultrasound ranging depends on assumptions about the average velocity of ultrasound in human tissue to locate reflectors in the ultrasound field. The elapsed time from pulse transmission to reception of the same pulse by the transducer allows for determining the location of a reflector in the ultrasound field

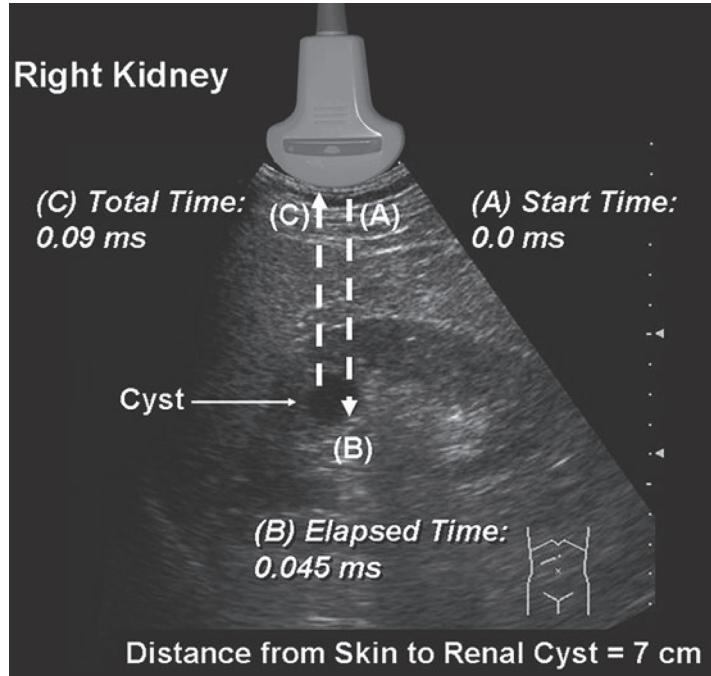
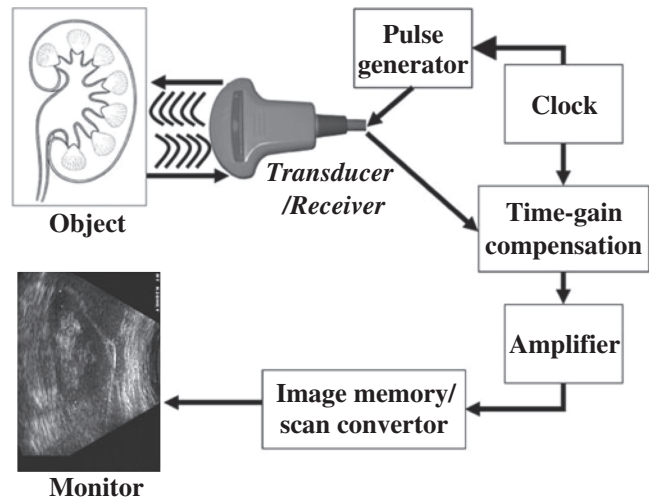


Fig. 2.7 Schematic depiction of the sequence of image production by an ultrasound device



Attenuation refers to a loss of kinetic energy as a sound wave interacts with tissues and fluids within the body [2]. Specific tissues have different potentials for attenuation. For example, water has an attenuation of 0.0 whereas kidney has an attenuation of 1.0 and muscle an attenuation of 3.3. Therefore, sound waves are much more

rapidly attenuated as they pass through muscle than as they pass through water (Fig. 2.8). (Attenuation is measured in dB/cm/MHz.)

The three most important mechanisms of attenuation are absorption, reflection, and scattering. Absorption occurs when the mechanical kinetic energy of a sound wave is converted to

Fig. 2.8 Attenuation of tissue.) (Adapted from Diagnostic Ultrasound, Third Ed., Vol 1). The attenuation of a tissue is a measure of how the energy of an ultrasound wave is dissipated by that tissue. The higher the attenuation value of a tissue, the more the sound wave is attenuated by passing through that tissue

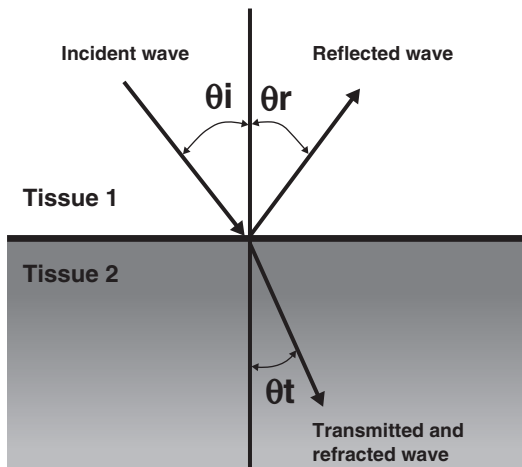
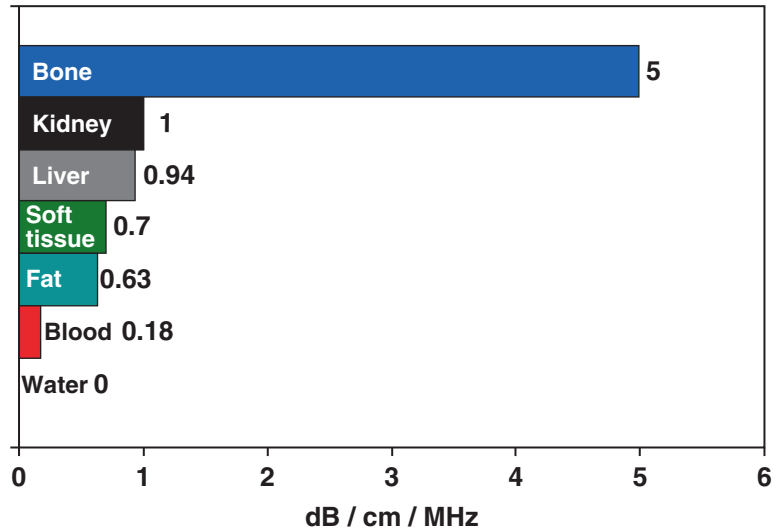


Fig. 2.9 A wave which strikes the interface between two tissues of differing impedance is usually partially reflected and partially transmitted with refraction. A portion of the wave is reflected at an angle (θ_r) equal to the angle of insonation (θ_i); a portion of the wave is transmitted at a refracted angle (θ_t) into the second tissue

heat within the tissue. Absorption is dependent on the frequency of the sound wave and the characteristics of the attenuating tissue. Higher frequency waves are more rapidly attenuated by absorption than lower frequency waves.

Since sound waves are progressively attenuated with distance traveled, deep structures in the body (e.g., kidney) are more difficult to image. Compensation for loss of acoustic energy by attenuation can be accomplished by the appropriate

use of gain settings (increasing the sensitivity of the transducer to returning sound waves) and selection of a lower frequency.

Refraction occurs when a sound wave encounters an interface between two tissues at any angle other than 90° . When the wave strikes the interface at an angle, a portion of the wave is reflected and a portion transmitted into the adjacent media. The direction of the transmitted wave is altered (refracted). This results in a loss of some information because the wave is not completely reflected back to the transducer, but also causes potential errors in registration of object location because of the refraction (change in direction) of the wave. The optimum angle of insonation to minimize attenuation by refraction is 90° (Fig. 2.9).

Reflection occurs when sound waves strike an object or an interface between unlike tissues or structures. If the object has a relatively large flat surface, it is called a specular reflector, and sound waves are reflected in a predictable way based on the angle of insonation. If a reflector is small or irregular, it is called a diffuse reflector. Diffuse reflectors produce **scattering** in a pattern which produces interference with waves from adjacent diffuse reflectors. The resulting pattern is called “speckle” and is characteristic of solid organs such as the testes and liver (Fig. 2.10).

When a sound wave travels from one tissue to another, a certain amount of energy is reflected at the interface between the tissues. The percentage

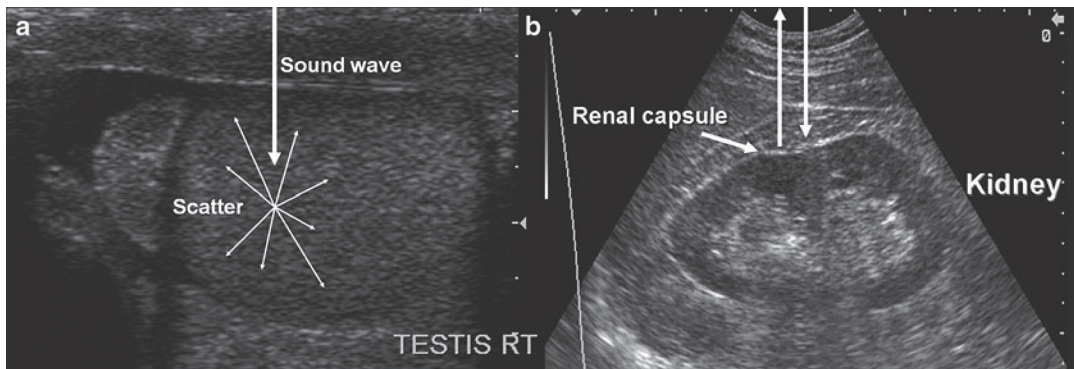


Fig. 2.10 (a) Demonstrates a diffuse reflector. In this image of the testis, small parenchymal structures scatter sound waves. The pattern of interference resulting from this scattering provides the familiar “speckled” pattern of testicular echo architecture. (b) Demonstrates a specular

reflector. A specular reflector reflects sound waves at an angle equal to the incident angle without producing a pattern of interference caused by scattering. In this image of the kidney, the capsule of the kidney serves as a specular reflector

Table 2.1 Impedance of tissue

Tissue	Density (kg/m ³)	Impedance (Rayles)
Air and other gases	1.2	0.0004
Fat tissue	952	1.38
Water and other clear liquids	1000	1.48
Kidney (average of soft tissue)	1060	1.63
Liver	1060	1.64
Muscle	1080	1.70
Bone and other calcified objects	1912	7.8

Adapted from Diagnostic Ultrasound, 3rd Ed, Vol. 1
 Impedance (Z) is a product of tissue density (p) and the velocity of that tissue (c). Impedance is defined by the formula: $Z \text{ (Rayles)} = p \text{ (kg/m}^3) \times c \text{ (m/s)}$

of energy reflected is a function of the difference in the **impedance** of the tissues. Impedance is a property of tissue related to its “stiffness” and the speed at which sound travels through the tissue [3]. If two adjacent tissues have a small difference in tissue impedance, very little energy will be reflected. The impedance difference between kidney (1.63) and liver (1.64) is very small so that if these tissues are immediately adjacent, it may be difficult to distinguish the interface between the two by ultrasound (Table 2.1).

Fat has a sufficient impedance difference from both kidney and liver that the borders of the two organs can be distinguished from the intervening fat (Fig. 2.11).

If the impedance differences between tissues are very high, complete reflection of sound waves may occur, resulting in acoustic shadowing (Fig. 2.12).

Artifacts

Sound waves are emitted from the transducer with a known amplitude, direction, and frequency. Interactions with tissues in the body result in alterations of these parameters. Returning sound waves are presumed to have undergone alterations according to the expected physical principles such as attenuation with distance and frequency shift based on the velocity and direction of objects they encountered. The timing of the returning echoes is based on the expected velocity of sound in human tissue. When these expectations are not met, it may lead to image representations and measurements which do not reflect actual physical conditions. These misrepresentations are known as “artifacts.” Artifacts, if correctly identified, can be used to aid in diagnosis.

Increased through transmission occurs when sound waves pass through tissue with less attenuation than occurs in the surrounding tissues. For example, when sound waves pass through a fluid-filled structure such as a renal cyst, the waves experience relatively little attenuation compared to that experienced in the

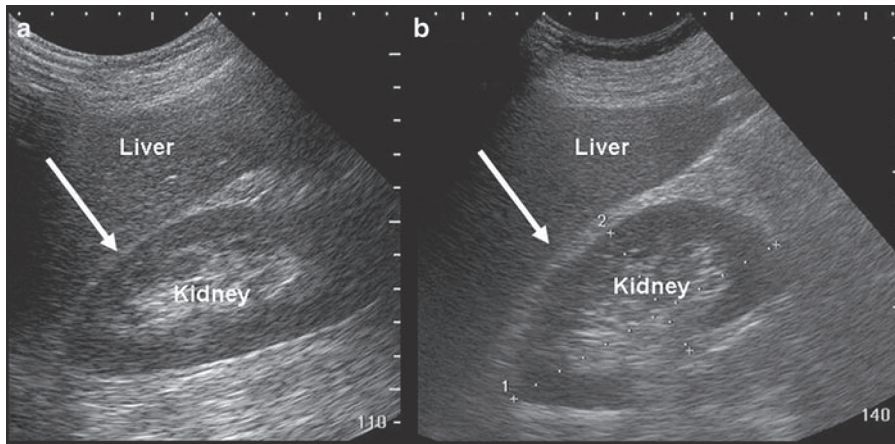


Fig. 2.11 Image (a) demonstrates that when kidney and liver are directly adjacent to each other, it is difficult to appreciate the boundary (arrow) between the capsules of the kidney and liver. Image (b) demonstrates that when fat

(which has a significantly lower impedance) is interposed, it is far easier to appreciate the boundary between liver capsule (arrow) and fat

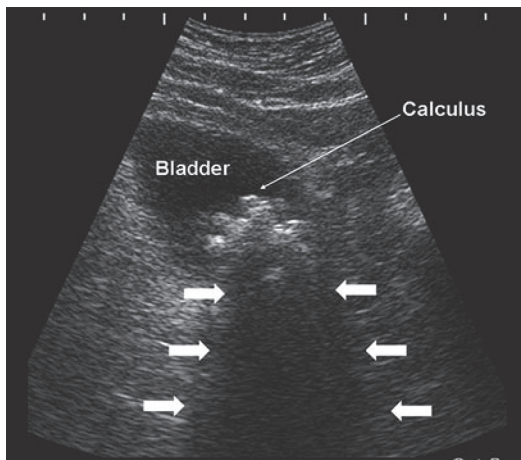


Fig. 2.12 In the urinary bladder, reflection of sound waves as the result of large impedance differences between urine and the bladder calculus (thin arrow). Acoustic shadowing results from nearly complete reflection of sound waves (arrows)

surrounding renal parenchyma. Thus when the waves reach the posterior wall of the cyst and the renal tissue beyond it, they are more energetic (have greater amplitude) than the adjacent waves. The returning echoes have significantly greater amplitude than waves returning through the renal parenchyma from the same region of the kidney. Therefore, the pixels associated with the region distal to the cyst are assigned a

greater “brightness.” The tissue appears hyperechoic compared to the adjacent renal tissue even though it is histologically identical (Fig. 2.13). This artifact can be overcome by changing the angle of insonation or adjusting the time-gain compensation settings.

Acoustic shadowing occurs when there is significant attenuation of sound waves at a tissue interface causing loss of information about other structures distal to that interface. This attenuation may occur on the basis of reflection or absorption, resulting in an “anechoic” or “hypoechoic” shadow. The significant attenuation or loss of the returning echoes from tissues distal to the interface may lead to incorrect conclusions about tissue in that region. For instance, when sound waves strike the interface between testicular tissue and a testicular calcification, there is a large impedance difference and significant attenuation and reflection occur. Information about the region distal to the interface is therefore lost or severely diminished (Fig. 2.14). Thus, in some cases spherical objects may appear as crescentic objects, and it may be difficult to obtain accurate measurements of such three-dimensional objects. Furthermore, fine detail in the region of the acoustic shadow may be obscured. The problems with acoustic shadowing are most appropriately overcome by changing the angle of insonation.

Fig. 2.13 Increased through transmission with hyperechogenicity (*arrow*) as the result of decreased attenuation by the fluid-filled cyst. This is an example of artifactual misrepresentation of tissue characteristics and must be recognized to avoid incorrect clinical conclusions

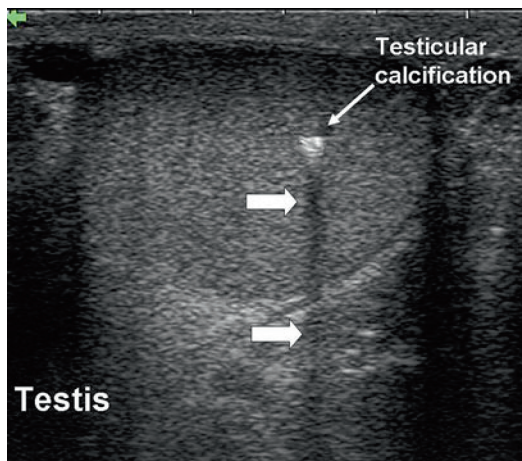
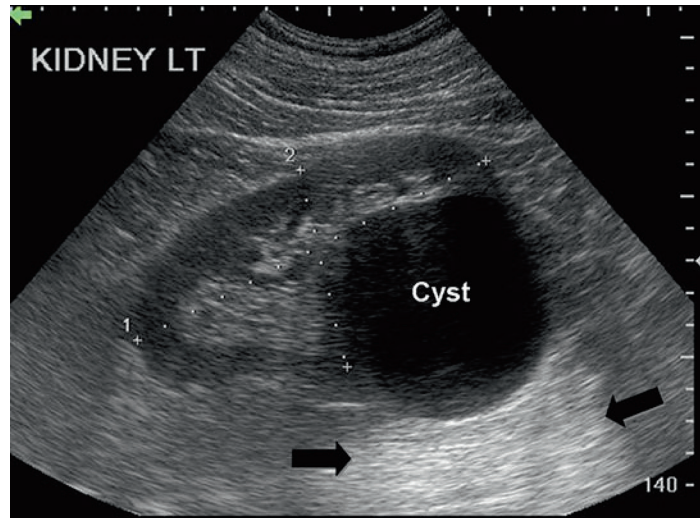


Fig. 2.14 Acoustic shadowing occurs distal to a calcification in the testis (*large arrows*). Information about testicular parenchymal architecture distal to the area of calcification is lost

Edging artifact occurs when sound waves strike a curved surface or an interface at a critical angle. A **critical angle** of insonation is one which results in propagation of the sound wave along the interface without significant reflection of the wave to the transducer. Thus, information distal to the interface is lost or severely diminished. This very common artifact in urology must be recognized and can, at times, be helpful. It is seen in many clinical situations but very commonly seen when imaging the testis. Edging artifacts often occur at the upper and lower pole of the

testis as the sound waves strike the rounded testicular poles at the critical angle. This artifact may help differentiate between the head of the epididymis and the upper pole of the testis. The edging artifact is also prominently seen on transrectal ultrasound, where the two rounded lobes of the prostate come together in the midline. This produces an artifact that appears to arise in the vicinity of the urethra and extend distally. Edging artifact may be seen in any situation where the incident wave strikes an interface at the critical angle (Fig. 2.15). Edging artifact may be overcome by changing the angle of insonation.

A **reverberation artifact** results when an ultrasound wave bounces back and forth (reverberates) between two or more reflective interfaces. When the sound wave strikes a reflector and returns to the transducer, an object is registered at that location. With the second transit of the sound wave, the ultrasound equipment interprets a second object that is twice as far away as the first. There is ongoing attenuation of the sound wave with each successive reverberation resulting in a slightly less intense image displayed on the screen. Therefore, echoes are produced which are spaced at equal intervals from the transducer but are progressively less intense (Fig. 2.16).

The reverberation artifact can also be seen in cases where the incident sound wave strikes a series of smaller reflective objects (such as the

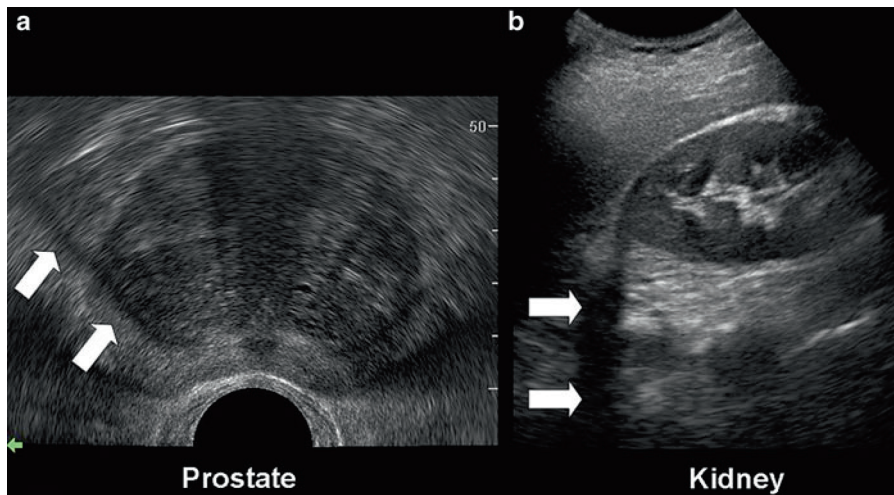


Fig. 2.15 Edging artifact (arrows) seen in this transverse image of the prostate is the result of reflection of the sound wave along the curved lateral surface of the transi-

tion zone (a). Edging artifact (arrows) caused by the rounded upper pole of the kidney (b)

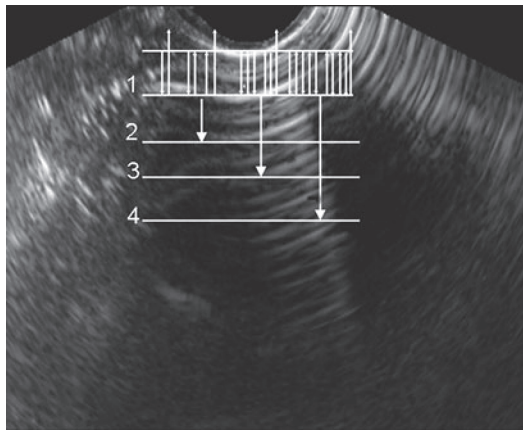


Fig. 2.16 A reverberation artifact occurs when a sound wave is repeatedly reflected between reflective surfaces. The resultant echo pattern is a collection of hyperechoic artifactual reflections distal to the structure with progressive attenuation of the sound wave

gas–fluid mixture in the small bowel) which results in multiple reflected sound waves of various angles and intensity (Fig. 2.17).

This familiar artifact may obscure important anatomic information and is frequently encountered during renal ultrasound. It may be overcome by changing the transducer location and the angle of insonation.

Modes of Ultrasound

Gray-Scale, B-Mode Ultrasound

Gray-scale, B-mode ultrasound (brightness mode) is the image produced by a transducer which sends out ultrasound waves in a carefully timed, sequential way (pulsed wave). The reflected waves are received by the transducer and interpreted for distance and amplitude. Time of travel is reflected by position on the image monitor and intensity by “brightness” of the corresponding pixel. Each sequential line-of-sight echo is displayed side by side and the entire image refreshed at 15–40 frames/s. This results in the illusion of continuous motion or “real-time” scanning. The intensity of the reflected sound waves may vary by a factor of 10^{12} or 120 dB. Although the transducer can respond to such extreme variations in intensity, most monitors or displays have an effective range of only 10^6 or 60 dB. Each of 512×512 or 512×640 pixels may display 2^8 or 256 shades of gray [3]. Most ultrasound units internally process and compress ultrasound data to allow it to be displayed on a standard monitor. Evaluation of

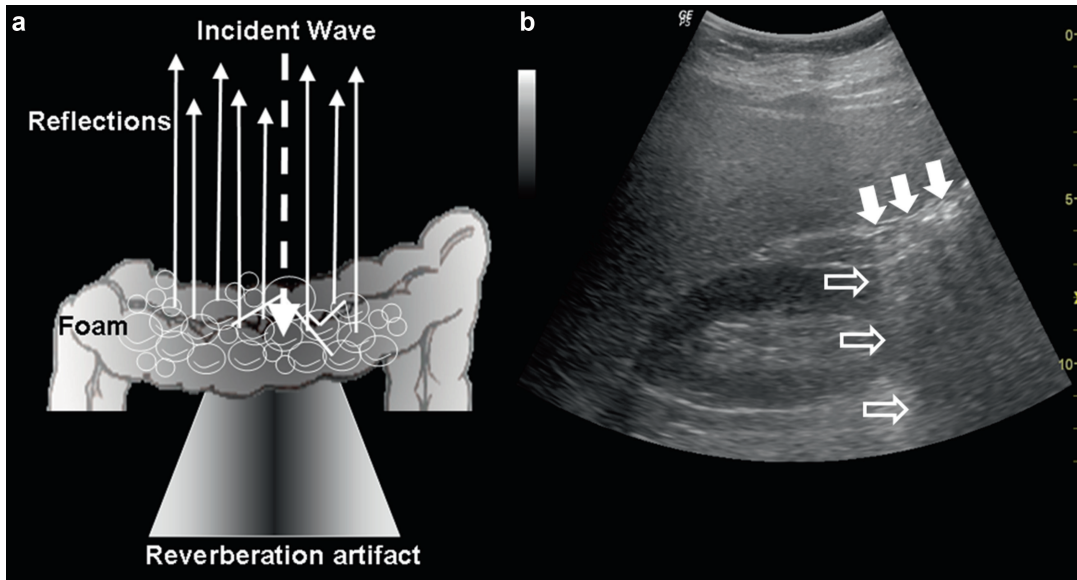


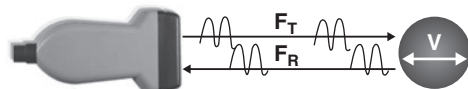
Fig. 2.17 (a) Reverberation artifact produced when sound waves strike a mixture of fluid and gas in the bowel. (b) This type of reverberation (multipath artifact) is char-

acterized by hyperechoic areas (*open arrow*) and distal attenuation of the incident wave (*closed arrows*)

Stationary target: $(F_R - F_T) = 0$



Target motion toward the transducer: $(F_R - F_T) > 0$



Target motion away from transducer: $(F_R - F_T) < 0$

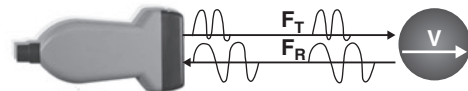


Fig. 2.18 Doppler effect. F_T is the transmitted frequency. When the F_T strikes a stationary object, the returning frequency F_R is equal to the F_T . When the F_T strikes a moving object, the F_R is “shifted” to a higher or lower frequency

gray-scale imaging requires the ability to recognize the normal patterns of echogenicity from anatomic structures. Variations from these expected patterns of echogenicity indicate disorders of anatomy or physiology or may represent artifacts.

Doppler Ultrasound

The Doppler ultrasound mode depends on the physical principle of frequency shift when sound waves strike a moving object. The basic principle of Doppler ultrasound is that sound waves of a certain frequency will be shifted or changed based on the direction and velocity of the moving object as well as the angle of insonation. This phenomenon allows for the characterization of motion, most commonly the motion of blood through vessels, but may also be useful for detecting the flow of urine.

The **Doppler Effect** is a shift in the frequency of the transmitted sound wave based on the velocity of the reflecting object that it strikes. If the reflecting object is stationary relative to the transducer, then the returning frequency will be equal to the transmitted frequency. However, if the echo-generating object is traveling toward the transducer, the returning frequency will be higher than the transmitted frequency. If the object generating the echo is traveling away from the transducer, then the reflected frequency will be lower than the transmitted frequency. This is known as the frequency shift, or Doppler shift (Fig. 2.18).

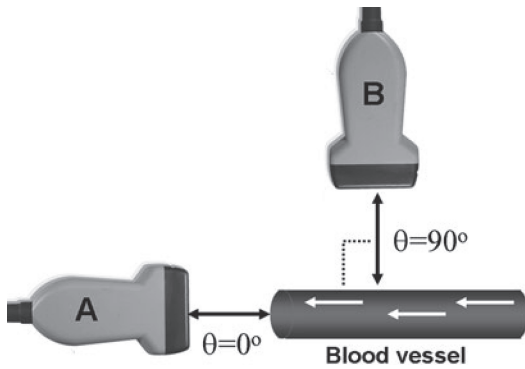


Fig. 2.19 (A) Maximum frequency shifts are detected when the transducer axis is parallel to the direction of motion. (B) No frequency shift is detected when the transducer axis is perpendicular to the direction of motion

The frequency shift of the transmitted wave is also dependent on the angle of the transducer relative to the object in motion. The maximum Doppler frequency shift occurs when the transducer is oriented directly on the axis of motion of the object being insonated. That is, when the transducer is oriented parallel (angle $\theta=0^\circ$) to the direction of motion, the shift is maximal. Conversely, when the transducer face is oriented perpendicular to the direction of motion (angle $\theta=90^\circ$), there will be no shift in Doppler frequency detected (Fig. 2.19).

An accurate calculation of velocity of flow depends on the angle θ between the transducer and the axis of motion of the object being insonated (Fig. 2.20).

Color Doppler ultrasonography allows for an evaluation of the velocity and direction of an object in motion. A color map may be applied to the direction. The most common color map uses blue for motion away from the transducer and red for motion toward the transducer (Fig. 2.21).

The velocity of motion is designated by the intensity of the color. The greater the velocity of the motion, the brighter is the color displayed. Color Doppler may be used to characterize blood flow in the kidney, testis, penis, and prostate. It also may be useful in the detection of “jets” of urine emerging from the urethral orifices. An accurate representation of flow characteristics requires attention to transducer orientation relative to the object in motion. Therefore, in most clinical circumstances the

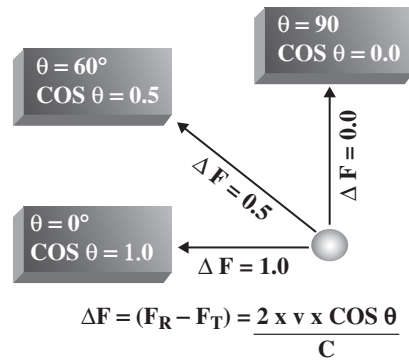


Fig. 2.20 Angle of insonation. The calculated velocity of an object using Doppler shift is dependent on the transducer angle (θ). If the transducer axis is perpendicular to the direction of flow (90°), then the cosine of θ is 0. Based on this formula for Doppler shift (ΔF), the detected frequency change would be 0. Adapted from Radiographics 1991;11:109–119

angle between the transducer and the direction of motion should be less than or equal to 60° (Fig. 2.22).

When it is not possible to achieve an angle of 60° or less by manipulation of the transducer, the beam may be “steered” electronically to help create the desired angle θ (Fig. 2.23).

Power Doppler ultrasonography is a mode which assigns the amplitude of frequency change to a color map. This does not permit evaluation of velocity or direction of flow but is less affected by backscattered waves. Power Doppler is therefore less angle dependent than color Doppler and is more sensitive for detecting flow [4].

When a sound wave strikes an object within the body, the sound wave is altered in a variety of ways including changes in frequency and changes in amplitude (Fig. 2.24).

While color Doppler assigns the changes in frequency to a color map, power Doppler assigns changes in integrated amplitude (or power) to a color map. It is possible to assign low-level backscattered information to a color which is unobtrusive on the color map, thereby allowing increased gain without interference from this backscattered information (Fig. 2.25). Power Doppler may be more sensitive than color Doppler for the detection of diminished flow [4].

The integrated amplitude (power) of the Doppler signal is signified by the brightness of the color.

Fig. 2.21 In this image of the radial artery, blood is flowing through the curved vessel from (A) to (C). Flow toward the transducer (A) is depicted in *red*. Flow in the middle of the vessel (B) is perpendicular to the transducer axis and produces no Doppler shift; thus, no color is assigned even though the velocity and intensity of flow are uniform through the vessel. Flow away from the transducer (C) is depicted in *blue*

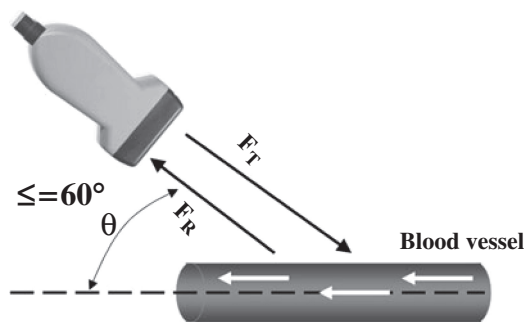
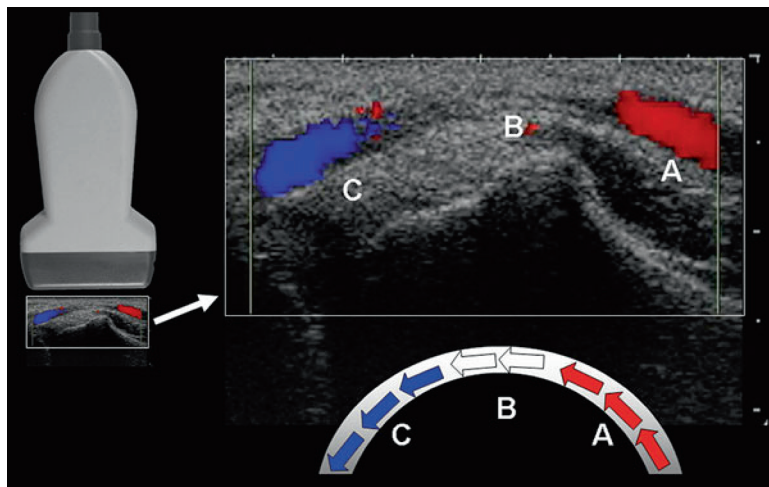


Fig. 2.22 The transducer angle should be $\leq 60^\circ$ relative to the axis of fluid motion to allow a more accurate calculation of velocity of flow

Because frequency shift is not displayed in standard power Doppler, the direction and velocity of flow are not indicated.

Color Doppler with spectral display is a mode which allows the simultaneous display of a color Doppler image and representation of flow as a wave form within a discrete area of interrogation. This mode is commonly used to evaluate the pattern and velocity of blood flow in the kidney and testis (Fig. 2.26).

The **spectral waveform** provides information about peripheral vascular resistance in the tissues. The most commonly used index of these velocities is the resistive index (Fig. 2.27).

The resistive index may be helpful in characterizing a number of clinical conditions including renal artery stenosis and urethral obstruction. Since

the velocity is represented on a scalar axis, it is necessary to set appropriate scalar limits to prevent artifacts. Therefore, it is necessary to know the expected velocity within vessels pertinent to urologic practice (Table 2.2). The clinical use of resistive index is described in subsequent chapters.

Artifacts Associated with Doppler Ultrasound

The **twinkle artifact** is produced when a sound wave encounters an interface which produces an energetic reflection of the sound wave. In ultrasound modes such as power and color Doppler, this can cause a distortion in the returning sound wave that gives the appearance of motion distal to that interface. The resulting Doppler signal appears as a trailing acoustic “shadow” of varying intensity and direction known as twinkle artifact. Although this artifact may be seen in a variety of clinical circumstances (e.g., twinkle artifact produced by the interaction of an ultrasound wave with a Foley catheter balloon in the bladder), it is most often helpful clinically in evaluating hyperechoic objects in the kidney. Stones often have a twinkle artifact (Fig. 2.28), whereas arcuate vessels and other hyperechoic structures in the kidney usually do not. Not all calcifications display the twinkle artifact. Calcifications of the renal artery and calcifica-

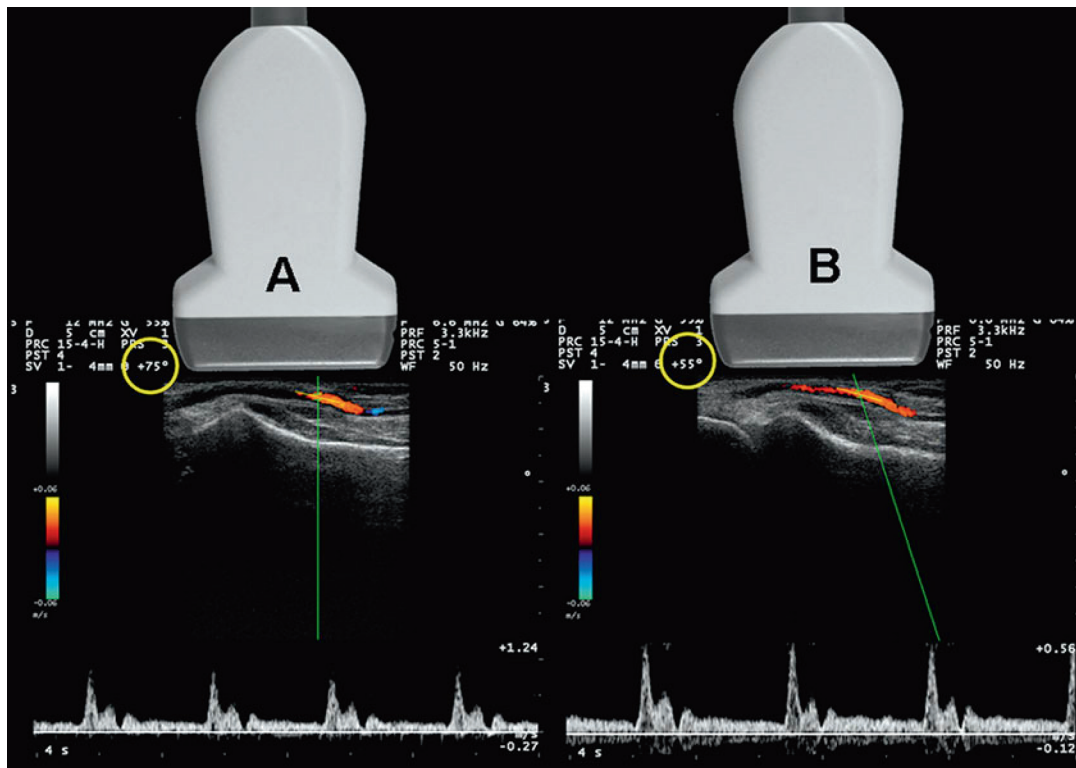


Fig. 2.23 Beam steering. In image (A), the angle of insonation is 75° (yellow circle) which is unfavorable for accurate velocity calculations. This is because the axis of the transducer is perpendicular to the vessel. In image (B)

the beam has been “steered” to produce an angle of 55° (yellow circle) without changing the physical position of the transducer. The resultant velocity calculation is more accurate at 55°

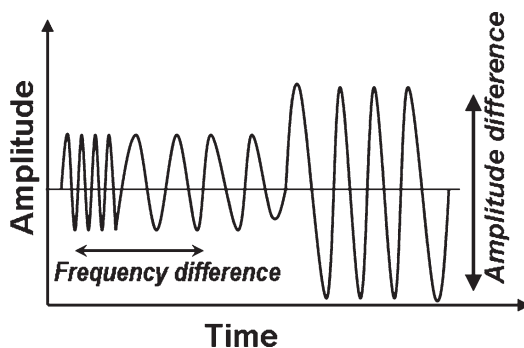


Fig. 2.24 Backscatter is defined as a combination of changes in frequency and amplitude which occur in the reflected sound wave of a primary frequency

tions within tumors and cysts may also produce the twinkle artifact [5].

Aliasing is an artifact which occurs when the ultrasound interrogation (determined by PRF) of an event occurs at a frequency which is insufficient

to accurately represent the event. When interrogation occurs at infrequent intervals, only portions of the actual event are depicted. Aliasing occurs when the interrogation frequency is less than twice the shifted Doppler frequency (Fig. 2.29).

Normal laminar unidirectional blood flow is depicted as a single color on color Doppler. Spectral Doppler shows a complete waveform (Fig. 2.30). During color Doppler scanning, aliasing is most commonly seen as apparent turbulence and change in direction of blood flow within a vessel. During spectral Doppler scanning, the aliasing phenomenon is seen as truncation of the systolic velocity peak with projection of the peak below the baseline (Fig. 2.31).

This artifact can be overcome by decreasing the frequency of the incident sound wave, increasing the angle of insonation (θ), or increasing the PRF.

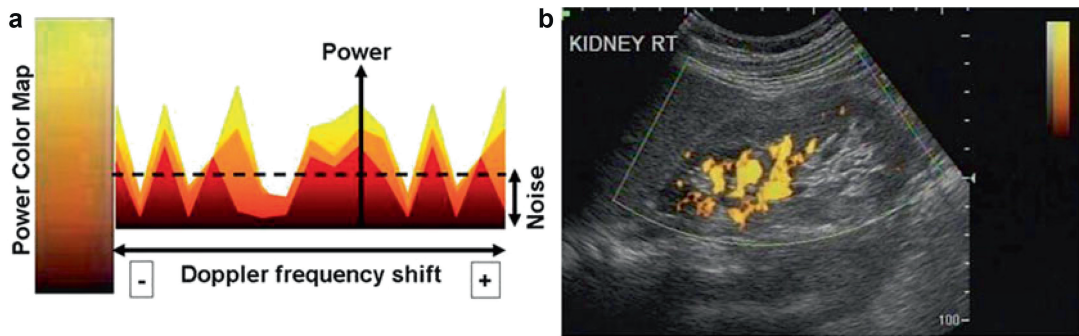
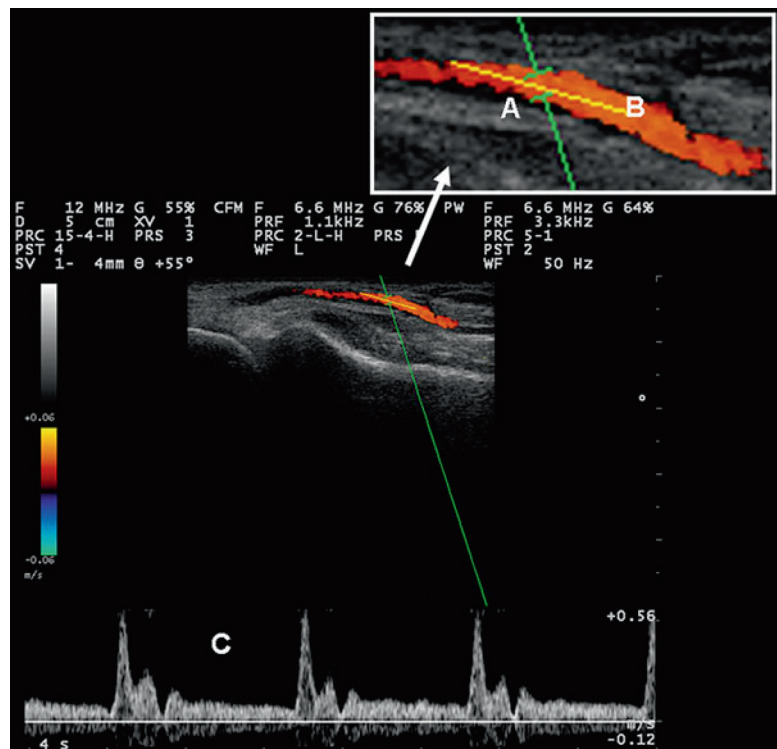


Fig. 2.25 (a) For power Doppler the intensity of color is related to changes in amplitude (power) rather than changes in frequency. (b) In this sagittal image of the kidney, power Doppler blood flow is demonstrated.

Note that the color map depicted to the upper right does not have a scale since quantitative measurement of velocity is not displayed with standard power Doppler

Fig. 2.26 In this example, the radial artery is shown in real-time gray-scale ultrasound with color Doppler overlay. The interrogation box or gate (A) is positioned over the vessel of interest. The gate should be positioned and sized to cover about 75% of the lumen of the vessel. The angle of insonation is indicated by marking the orientation of the vessel with a cursor (B). The velocity of the flow within the vessel is depicted quantitatively in the spectral display (C)



Harmonic Scanning

Harmonic scanning makes use of aberrations related to the nonlinear propagation of sound waves within tissue. These asymmetrically propagated waves generate fewer harmonics but those which are generated have greater amplitudes (Fig. 2.32).

Since these harmonics are less subject to scattering associated with the incident wave, there is less noise associated with the signal. By selectively displaying the harmonic frequencies which are produced within the body and reflected to the transducer, it is possible to produce an image with less artifact and greater resolution.

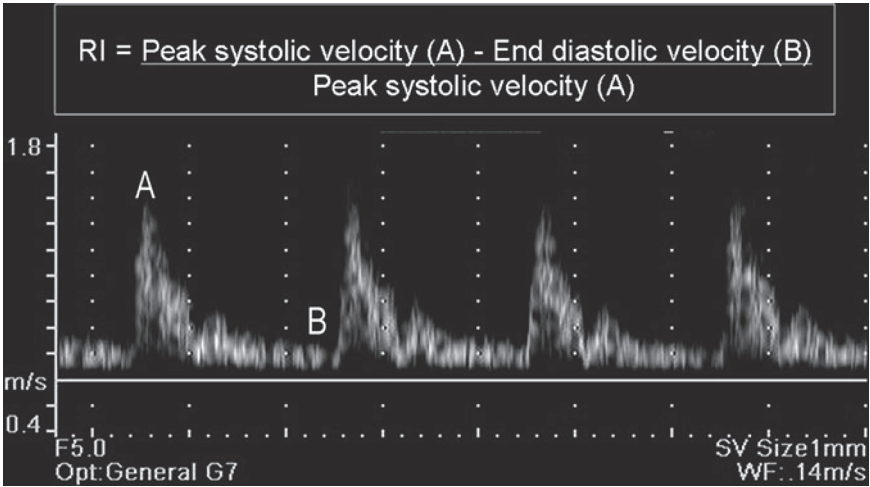


Fig. 2.27 The resistive index (RI) is the peak systolic velocity (A) minus the end-diastolic velocity (B) over the peak systolic velocity (A)

Table 2.2 Expected velocity in urologic vessels

Vessel	Velocity
Penile artery	>35 cm/s (after vasodilators) [10]
Renal artery	<100 cm/s [11]
Scrotal capsular artery	5–14 cm/s [12]

The measured velocity will depend on a variety of physiologic and anatomic variants

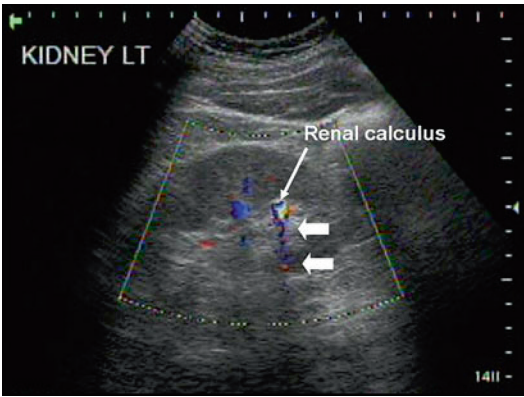


Fig. 2.28 Twinkle artifact. The effect produced by the interaction of sound waves at an interface with high impedance differences (in this case a renal calculus) which produces an artifact suggesting turbulent motion (large arrows)

Spatial compounding is a scanning mode whereby the direction of insonation is sequentially altered electronically to produce a composite image. This technique reduces the amount of artifact and noise producing a scan of better clarity [6] (Fig. 2.33).

Three-dimensional (3D) scanning produces a series of images (data set) which can then be manipulated to generate additional views of the anatomy in question. 3D rendering may be important in procedural planning and precise volumetric assessments [7]. 3D scanning may allow the recognition of some tissue patterns which would otherwise be inapparent on two-dimensional scanning [8].

Contrast Agents in Ultrasound

Intravenous compounds containing microbubbles have been used for enhancing the echogenicity of blood and tissue. Microbubbles are distributed in the vascular system and create strong echoes with harmonics when struck by sound waves. The bubbles themselves are rapidly degraded by the interaction with the sound waves. Machine settings producing a low mechanical index (see Chap. 4) are desirable to reduce destruction of the microbubbles. Contrast agents may be useful

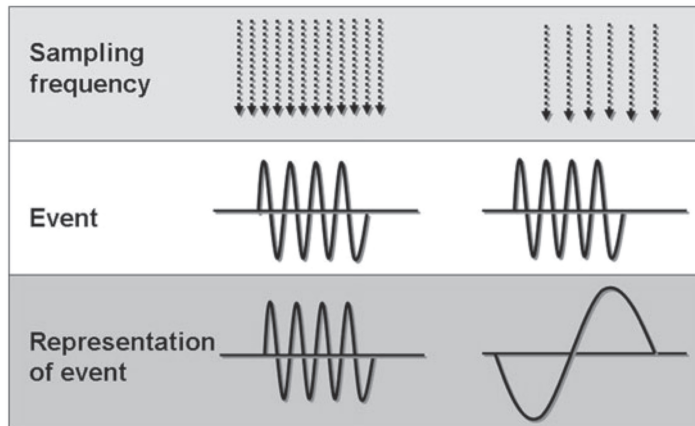


Fig. 2.29 Aliasing. In this illustration where a sine wave is the real-time event and the vertical *arrows* in the *top* panel represent the frequency of interrogation, we see that frequent interrogation produces an accurate representation of the event. An accurate depiction of an ultrasound event must meet the condition: $f_s \geq 2b$, where f_s is the

sampling frequency and $2b$ is the highest frequency in the event. This is known as the Nyquist limit. Less frequent sampling (on the right) results in an incorrect interpretation of the event. (Diagram adapted from Diagnostic Ultrasound, 3rd Ed., Figs. 1–40, p. 33)

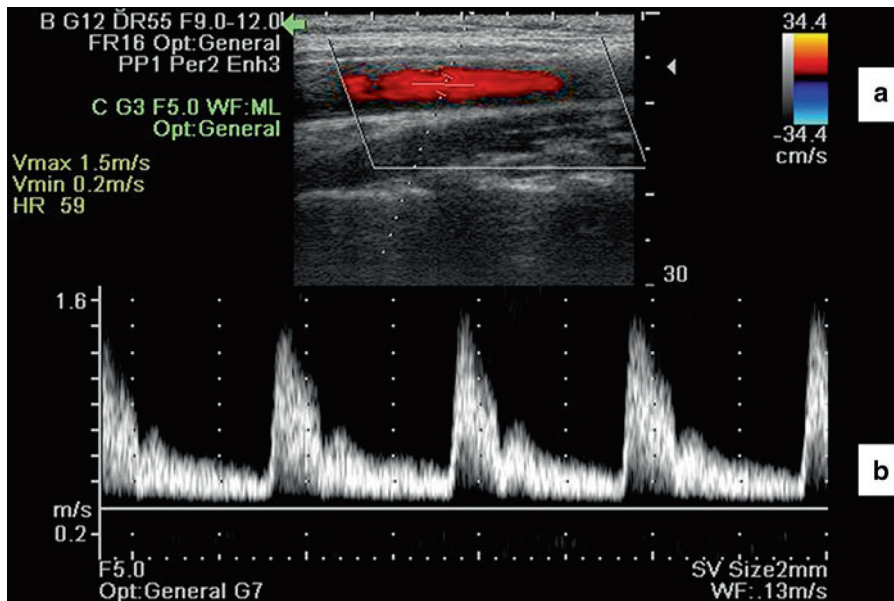


Fig. 2.30 Spectral Doppler. (a) Blood flow appears unidirectional on color mapping in this color Doppler image with spectral flow analysis. (b) The waveform is accurately depicted on spectral analysis

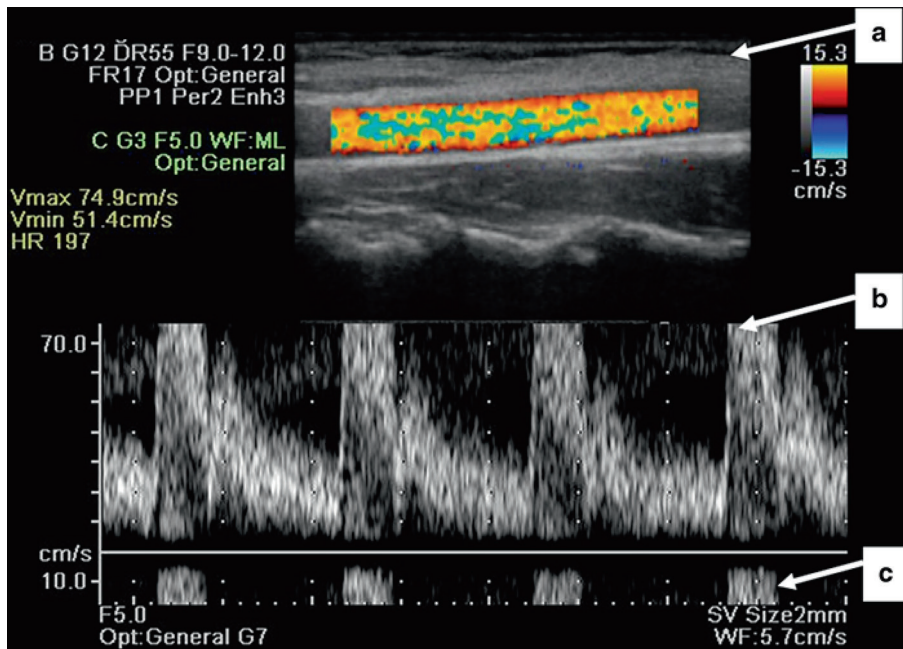


Fig. 2.31 In this color Doppler ultrasound with spectral flow, aliasing is demonstrated by apparent changes in velocity and direction on the color map assigned to the

vessel (a). Aliasing of the spectral waveform is seen as truncation of the peak systolic velocity (b) with projection of the peak below the baseline (c)

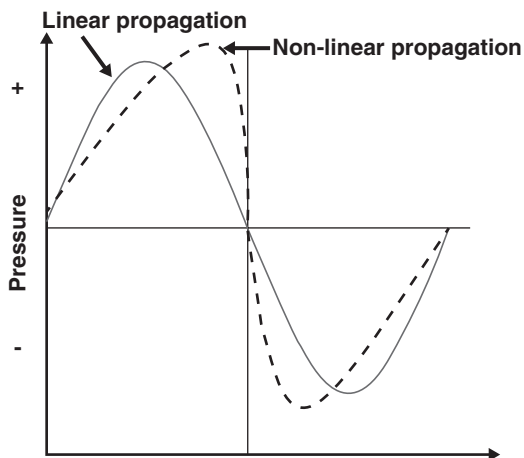


Fig. 2.32 Nonlinear propagation of sound waves in tissue results in fewer but more energetic harmonics which may be selectively evaluated in the returning echo

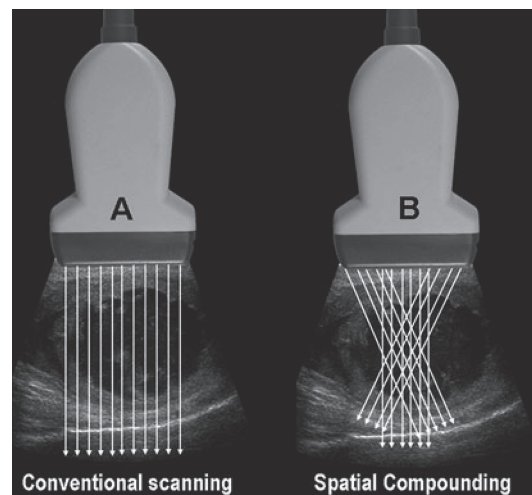


Fig. 2.33 Spatial compounding results in a composite image by combining data from multiple scanning angles produced by automated beam steering. The resulting image is more detailed with less artifact

in prostatic ultrasound by enhancing the ability to recognize areas of increased vascularity [9]. The use of intravenous ultrasound contrast agents is considered investigational but has shown promise in a number of urologic scanning situations. The ability to demonstrate vascular enhancement without exposure to potential toxic contrast agents or ionizing radiation makes contrast-enhanced ultrasound a promising urologic imaging modality.

References

1. Mason WP. Piezoelectricity, its history and applications. *J Acoust Soc Am*. 1981;70(6):1561–6.
2. Rumack CM, Wilson SR, Charboneau JW. Diagnostic ultrasound. 3rd ed. St. Louis: Mosby; 2005. p. 8.
3. Rumack CM, Wilson SR, Charboneau JW. Diagnostic ultrasound, vol. 3. St. Louis: Mosby; 2005. p. 12.
4. Rubin JM, Bude RO, Carson PL, et al. Power Doppler US: a potentially useful alternative to mean frequency-based color Doppler US. *Radiology*. 1994; 190:853–6.
5. Kim HC, Yang DM, Jin W, Ryu JK, Shin HC. Color Doppler twinkling artifacts in various conditions during abdominal and pelvic sonography. *J Ultrasound Med*. 2010;29:621–32.
6. Merritt CR. Technology update. *Radiol Clin North Am*. 2001;39:385–97.
7. Ghani KR, Pilcher J, Patel U, et al. Three-dimensional ultrasound reconstruction of the pelvicaliceal system: an in-vitro study. *World J Urol*. 2008;26:493–8.
8. Mitterberger M, Pinggera GM, Pallwein L, et al. The value of three-dimensional transrectal ultrasonography in staging prostate cancer. *BJU Int*. 2007;100:47–50.
9. Mitterberger M, Pinggera GM, Horninger W, et al. Comparison of contrast enhanced color Doppler targeted biopsy to conventional systematic biopsy: impact on Gleason score. *J Urol*. 2007;178:464–8.
10. Rifkin MD, Cochlin DL. Imaging of the scrotum & penis. London: Martin Dunitz; 2002. p. 276.
11. Zucchelli PC. Hypertension and atherosclerotic renal artery stenosis: diagnostic approach. *J Am Soc Nephrol*. 2002;13:S184–6.
12. Bluth EI, Benson CB, Ralls PW. Ultrasonography in vascular diseases: a practical approach to clinical problems. New York: Thieme Medical; 2008. p. 87.

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