Introduction to Ultrasound

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Basics

INTRODUCTION

One begins, of course, with the basics. In the field of diagnostic ultrasound, the basics are physics, instrumentation, and safety, and all three of these subjects are addressed in this section. Practitioners of diagnostic sonography, by and large, have limited backgrounds in physics and engineering, and for this reason the two chapters that make up this section are *really* basic. The goal here is to convey fundamental concepts about diagnostic ultrasound in such a way that they can be understood by virtually anyone, including the authors. More sophisticated readers may be disappointed in the content of these chapters, but these individuals will be pleased to know that numerous highly technical sources exist from which they can obtain additional information.

Chapter 1
Basic Ultrasound Physics and Instrumentation

William J. Zwiebel

Ultrasound physics and instrumentation[1-5] are reviewed briefly in this chapter, principally in the format of captioned illustrations. This is an unusual approach for teaching ultrasound physics, but I feel that this method conveys the concepts of ultrasound physics well and is relatively painless.

Physicists or engineers reading this chapter may be horrified to find that only a few mathematical formulas are presented! I have avoided the use of mathematical formulas because the majority of medical personnel have not been trained to think in mathematical terms. A mathematical format, therefore, can make ultrasound physics seem more complicated than necessary.

DEFINITIONS

A few definitions are in order at the outset. Please review these briefly now and return to them as needed as you proceed through this and the following chapter.

B-mode—Abbreviation for "brightness modulation mode," which is the basis for all ultrasound images. Echoes are converted into bright dots that vary in intensity (are modulated) according to the strength of the echo.

Farfield—The portion of the ultrasound image distant from the transducer.*

Frame rate—The rate at which the image on an ultrasound display screen (television monitor) is renewed. The frame rate must exceed 20 frames per second to avoid image "flickering."

Frequency—The number of ultrasound waves per second (Fig. 1–1).

Gray scale—The display of various levels of echo brightness in shades of gray; as opposed to a "bistable" display in which only black and white are shown.

Nearfield—The portion of the ultrasound image near the transducer.*

Pulse echo sonography—Ultrasound technique using a single transducer to send short bursts (or pulses) of ultrasound into the body and alternately "listen" for echoes.

Pulse repetition frequency—The number of ultrasound pulses sent into the body per second.

RADAR—Abbreviation for "radio detection and ranging."

Range—Technical synonym for "distance."

Real time—Abbreviated term meaning "in real time"; that is, movement is depicted on the display screen as it occurs, without an appreciable delay.

Scanhead—The composite of the ultrasound crystal, its electrical attachments, and its housing. In short, the entire ultrasound sending and receiving device that comes into contact with the patient.

Scanner—The entire ultrasound instrument, including the scanhead.

SONAR—Abbreviation for "sound navigation and ranging."

Sonography—The process of generating images with ultrasound (analogous to the term "photography").

Transducer—A device that converts one form of energy into another. The ultrasound crystal is the heart of the transducer. It converts electrical energy to ultrasound (mechanical energy) and vice versa.

*These terms are defined here in a general sense. Nearfield and farfield actually refer to specific portions of an ultrasound beam, which is beyond the scope of this chapter.

Two-dimensional image—An image that has width and height. Photographs and television images are two-dimensional.

Ultrasound—Sound that exceeds a frequency level of 2000 cycles per second (2 kilohertz).[1] For medical diagnosis, ultrasound frequencies of 2.5 to 10 million cycles per second (2–10 megahertz) are used commonly.

Wavelength—The distance encompassed by each ultrasound wave (Fig. 1–1).

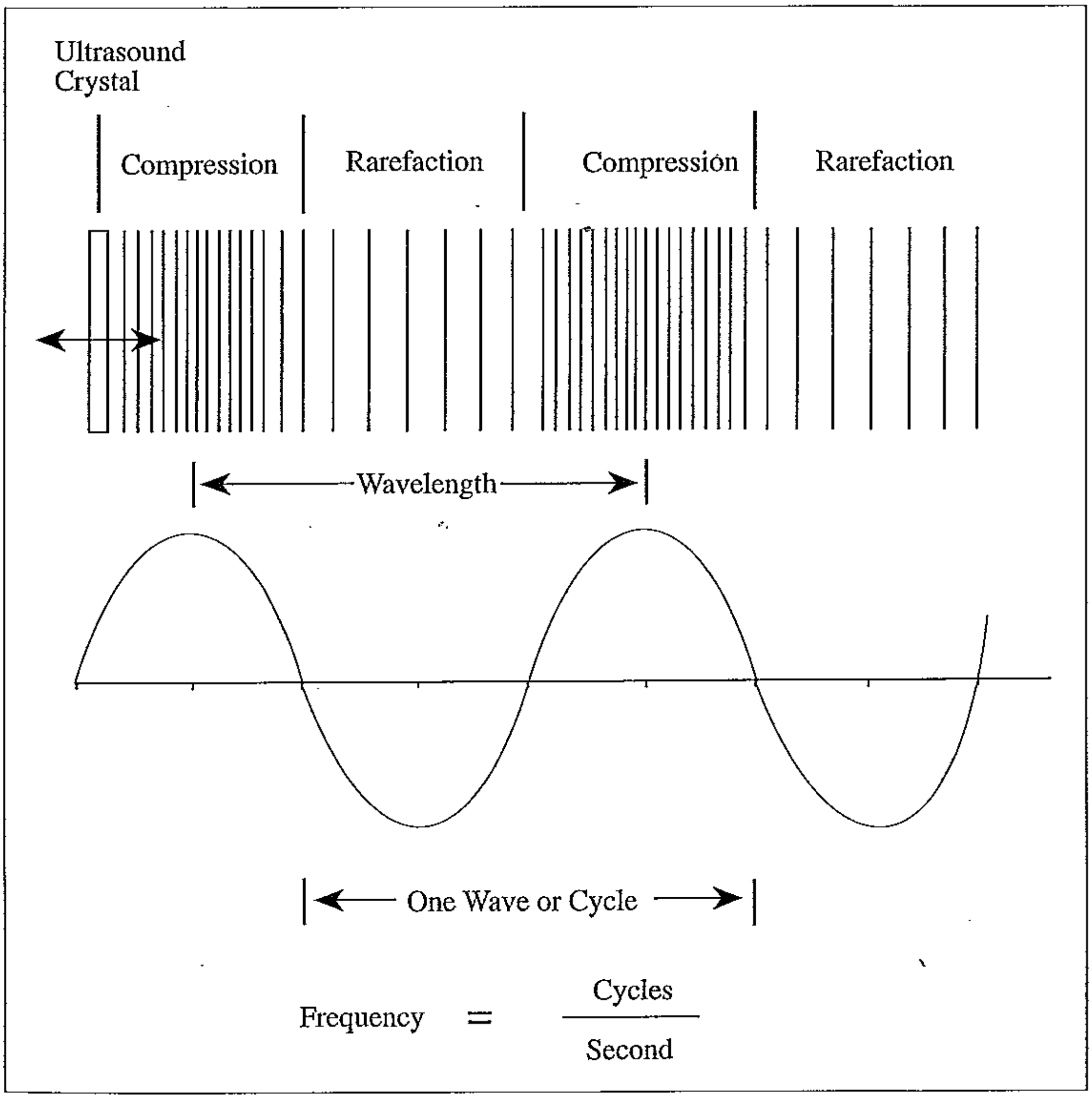


Figure 1-1—Illustration of an Ultrasound Wave. As the ultrasound crystal (upper left) vibrates, it sends an ultrasound *wave* into the body that consists of alternating *compression* and *rarefaction* zones (that is, the tissues are alternately compressed and stretched as the wave passes through).

The ultrasound wave can be illustrated as a graph, as seen in the midportion of this figure. Each complete "wave" or cycle contains one compression and one rarefaction zone, and each wave is represented on the graph as one peak and one valley. The *wavelength* is the distance between consecutive peaks (as shown here), consecutive valleys, or consecutive crossings of the baseline.

The *frequency* is the number of waves (or cycles) per second. The term *hertz** is commonly used when referring to frequency. One cycle per second equals 1 hertz (Hz); therefore, 1000 cycles per second equals a kilohertz (KHz) and 1,000,000 cycles per second equals a megahertz (MHz). When we say that a certain ultrasound transducer operates at 5 megahertz, we are using shorthand meaning that it operates at 5 million cycles per second.

*This term honors a famous German physicist named Hertz (not the car rental company).

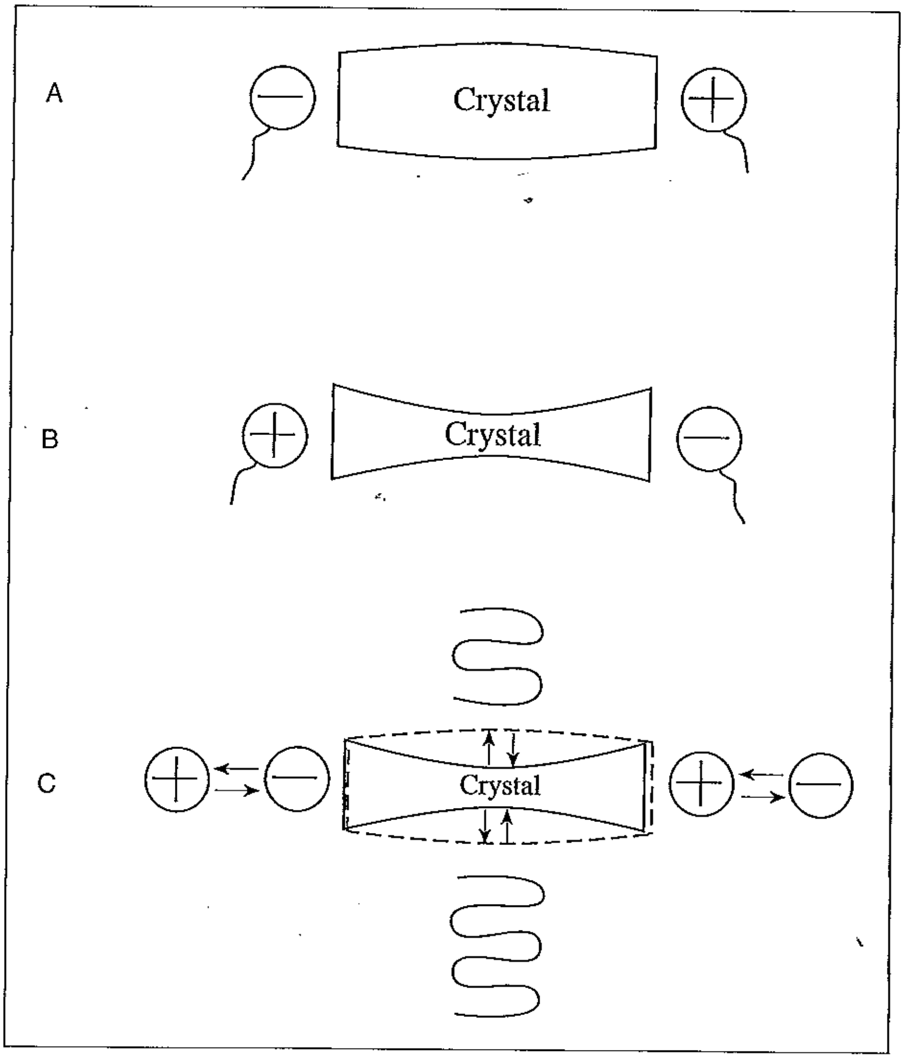


Figure 1-2—Ultrasound Production. Ultrasound is produced by the vibration of a synthetic crystal that possesses *piezoelectric* properties. When an electrical potential is applied to a piezoelectric crystal, it either expands (**A**) or contracts (**B**), depending on the polarity of the electrical connections. Ultrasound is generated when a rapidly alternating electrical potential causes the crystal to vibrate (**C**).

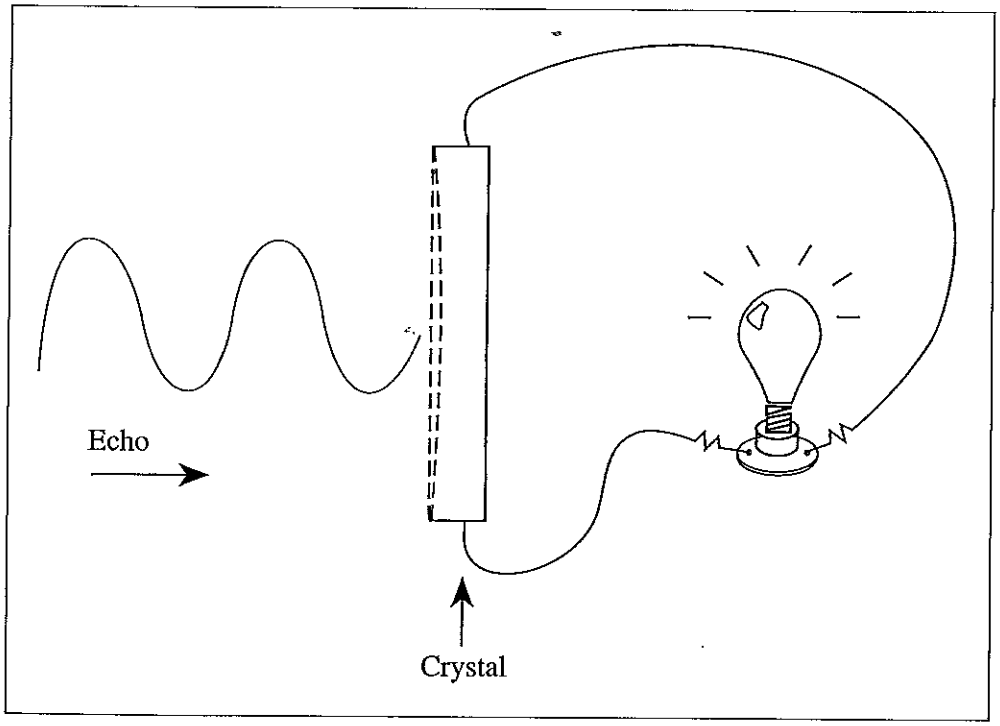


Figure 1-3—Ultrasound Reception. Ultrasound "echoes" reflected from objects within the body are detected by the same piezoelectric crystal that produced the ultrasound waves. The returning waves deform the crystal, generating a minute electrical potential that is sensed by the instrument and recorded electronically. This electrical potential is exceedingly weak and could not possibly light a light bulb, as facetiously shown here.

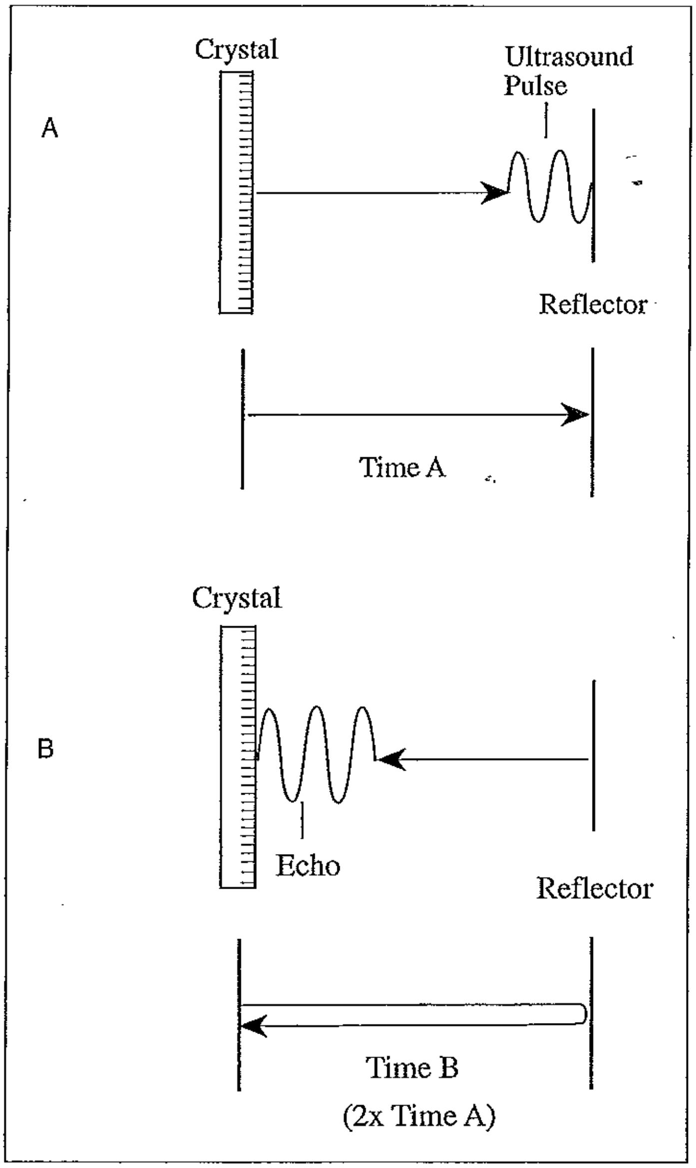


Figure 1-4—Pulse-Echo Sonography. For most medical applications, the piezoelectric crystal is stimulated electrically for only very short periods and produces, therefore, brief "pulses" of ultrasound (**A**). The ultrasound crystal then "listens" for echoes from structures within the body (**B**). Ultrasound pulses are very short; and the listening period is about 1000 times longer than the sending period.

For a businessperson, time is money, but for an ultrasound instrument, time is distance; that is to say, distance is measured as the *transit time* of the ultrasound pulse—from the ultrasound crystal to a reflector and then back to the crystal (Time B in part **B**). The *longer* the transit time, the *greater* the distance from the crystal to the reflector. To calculate distance, the instrument assumes that ultrasound travels through soft tissues at a uniform velocity of 1540 meters per second.[2] Transmission velocity actually is not uniform for all soft tissues, but image distortion resulting from velocity variation usually is insignificant.

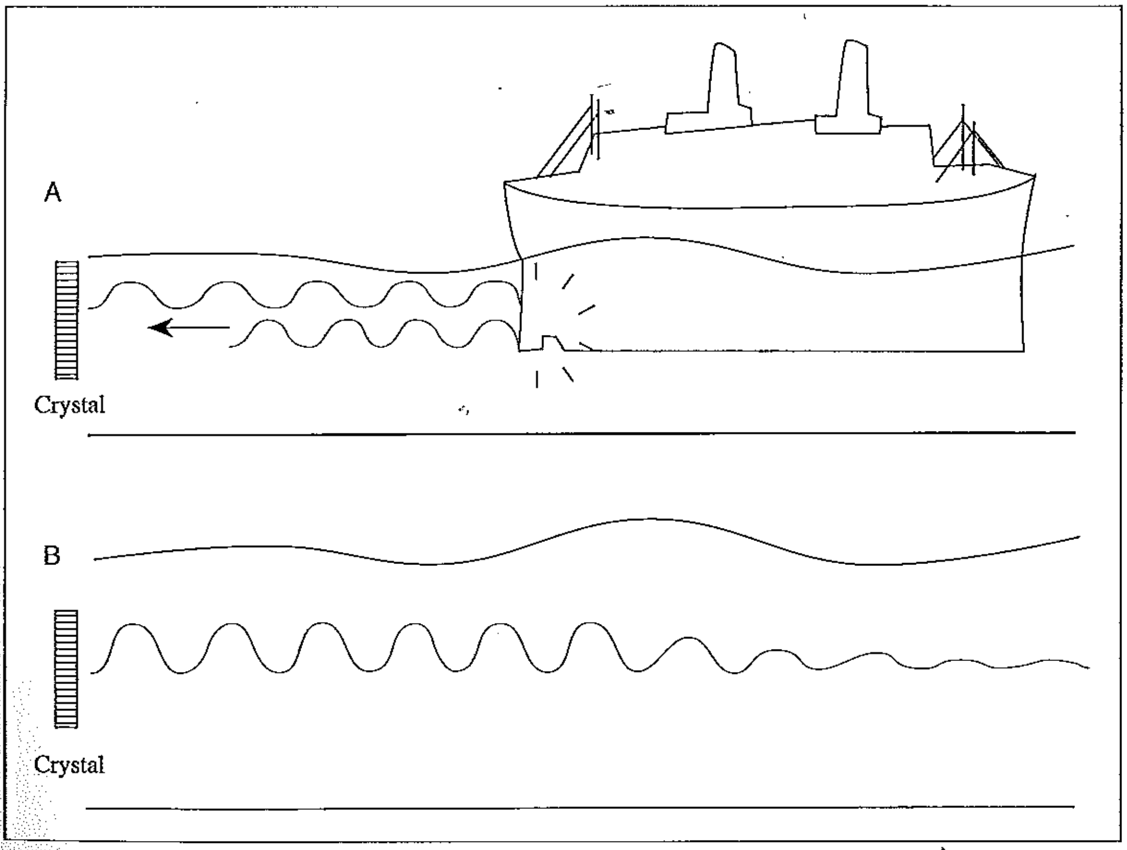


Figure 1-5—Reflecting Interfaces. Ultrasound is reflected only at boundaries, or *interfaces,* between two materials that have *different* acoustic properties. **(A)** Ultrasound pulses transmitted throughout the ocean are reflected when they strike a ship, since the steel from which the ship is made has vastly different acoustic properties than the adjacent water. **(B)** If ultrasound pulses traveling through the ocean do not encounter a ship, or anything other than water, they will travel outward until they fade from existence (from frictional forces), and they never will return to the transducer.

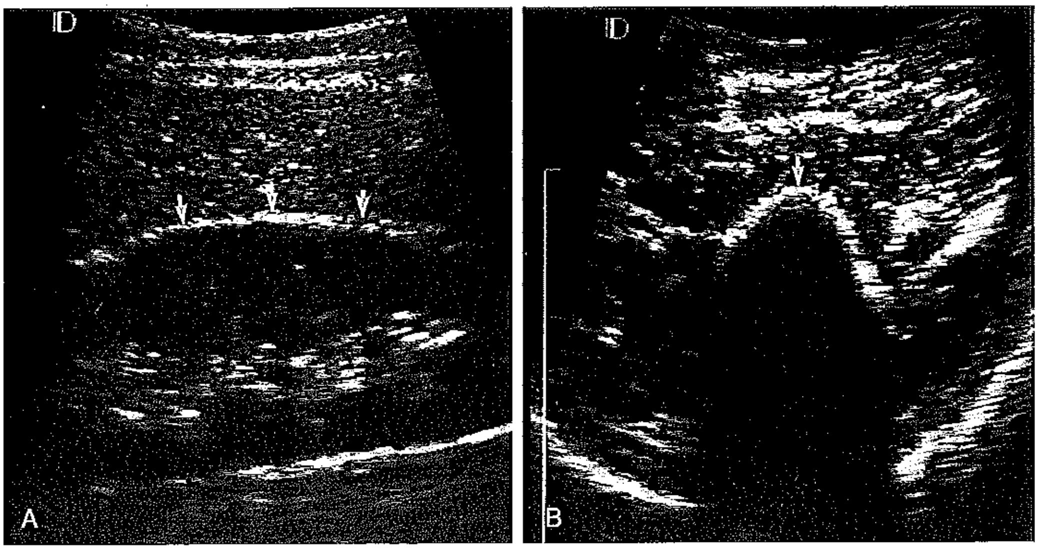


Figure 1-6—Echo Strength, Factor 1. A major factor affecting echo strength is the *degree* to which the acoustic properties differ for materials making up a reflecting interface (or boundary). This difference is called the *acoustic impedance mismatch.* The greater the mismatch, the greater the reflection, and the less ultrasound penetrates the interface. For instance, as shown in **(A)**, strong echoes are generated at the boundary between the liver and perinephric fat (arrows) because the acoustic properties of these tissues are different. The acoustic impedance mismatch is not so great as to block ultrasound transmission entirely, however, and deeper structures are clearly visible. **(B)** In contrast, virtually all the ultrasound energy is reflected at a bone/soft tissue boundary (arrow) because the acoustic mismatch between bone and soft tissue is very great. An "acoustic shadow" occurs distal (deep) to the bone since virtually all the ultrasound is reflected.

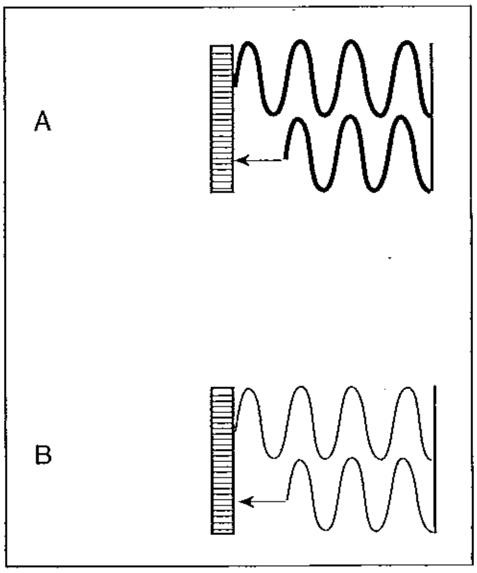


Figure 1-7—Echo Strength, Factor 2. The strength of an ultrasound echo also is governed by the inherent *strength of the ultrasound beam.* A powerful ultrasound pulse **(A)** generates stronger reflections than a weak pulse **(B)**. It might seem that the strongest possible ultrasound beam should be used to enhance image quality, but three factors necessitate the restriction of beam strength: (1) if echoes are too strong they will "overwhelm" the ultrasound receiver, causing "white out"; (2) excessive beam strength (ultrasound power) can cause tissue damage; and (3) excessive beam strength could generate heat perceptible by the patient.

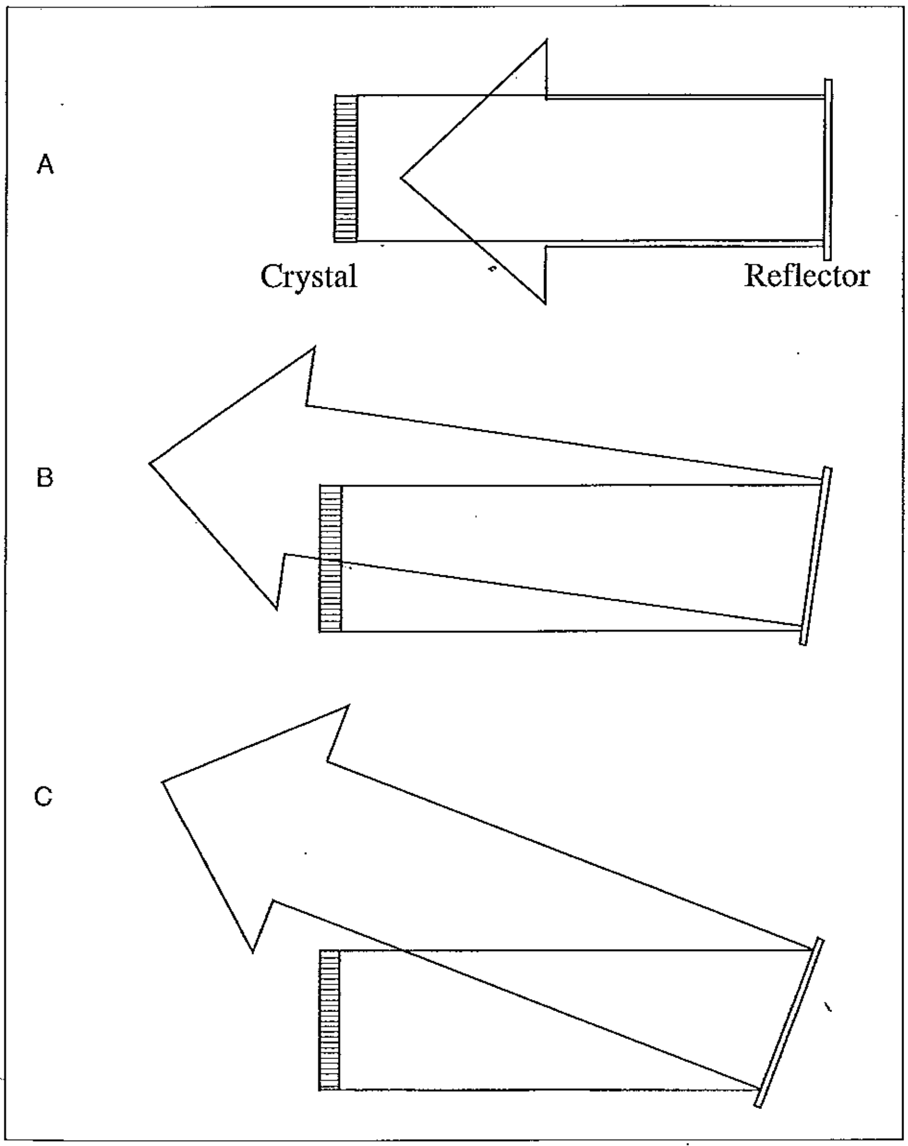


Figure 1-8—Echo Strength, Factor 3. The strength of a reflection also is affected by the angle at which the ultrasound beam strikes a reflecting interface. In technical jargon, this angle is called the *angle of incidence.* As is the case with light, the angle of incidence of an ultrasound beam equals the angle of reflection. **(A)** Maximum reflection back to the transducer, hence the strongest echo, occurs when the angle of incidence is 90°; that is, when the reflector is perpendicular to the ultrasound beam. **(B)** The strength of the echo wanes as the angle of incidence decreases. **(C)** At some point, the ultrasound beam is deflected away from the crystal and no echo is recorded.

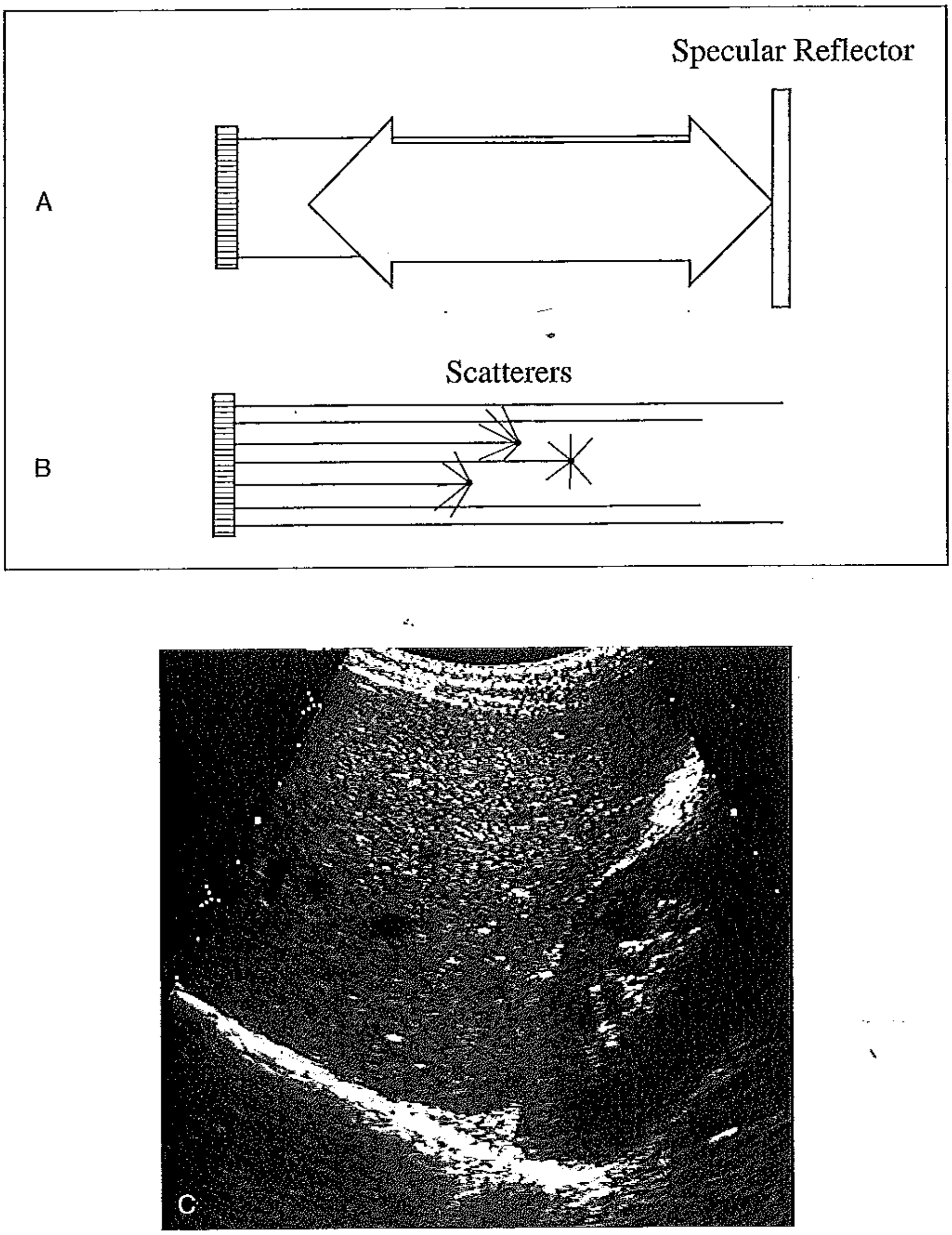


Figure 1-9—Echo Strength, Factor 4. The size of the reflector significantly affects the strength of an ultrasound reflection. Reflecting interfaces fall into two classes: **(A)** *specular reflectors,* which are large compared with the ultrasound beam and produce high-intensity, unidirectional reflections, and **(B)** *scatterers,* or tiny, punctate reflectors that scatter a small portion of the ultrasound beam in virtually all directions. (A chip in an automobile windshield acts as a scatterer when struck by the beam of an oncoming headlight.) **(C)** Specular reflectors provide the broad outlines of organs, whereas scatterers provide the "sonographic texture" within the organs. This texture actually arises from a phenomenon called speckle, as discussed in Chapter 9. Early ultrasound instruments could display only strong specular reflections; therefore, they could display only the outlines of large anatomic structures.

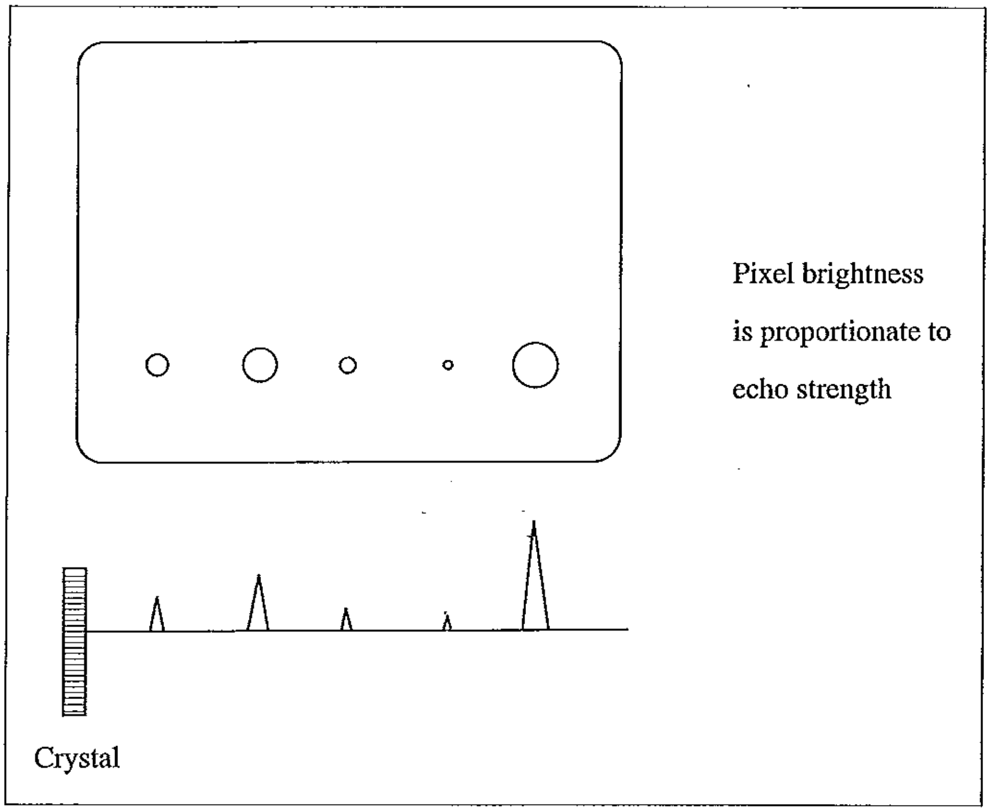


Figure 1-10—The B-Mode Display. All standard ultrasound images are brightness modulated, or "B-mode" images. Bright dots or *pixels* (short for picture elements) make up the picture, and the brightness of each pixel (illustrated here by the size of the circle) is modulated (adjusted) in proportion to the strength of each echo (illustrated by the height of the spikes).

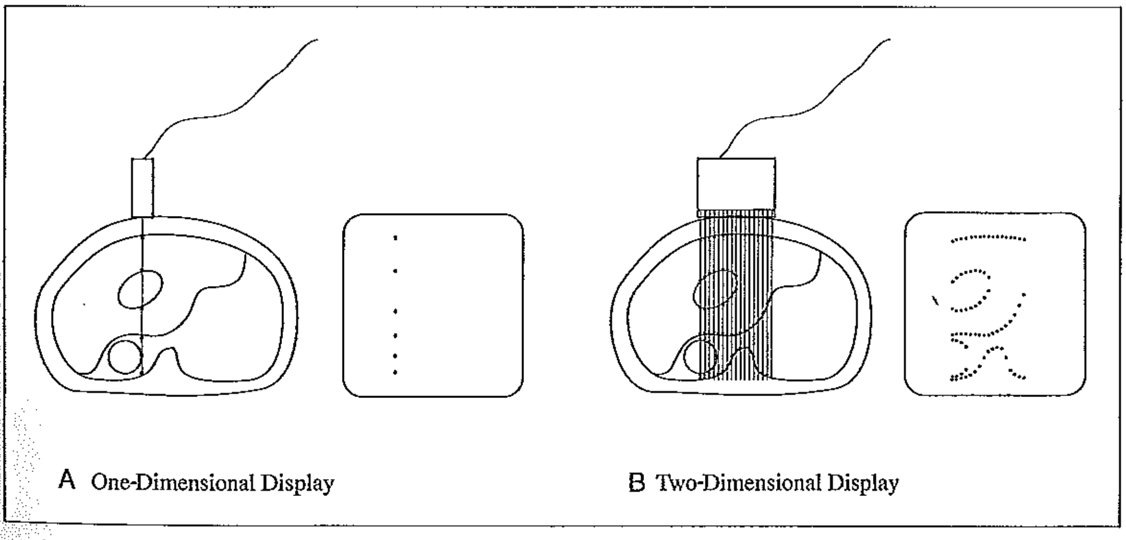


Figure 1-11—One- and Two-Dimensional Displays. (A) The B-mode display from a single ultrasound crystal is a one-dimensional series of bright dots. This is the ultimate "ice-pick" or "searchlight" view of the world. (B) Multiple "ice-pick" views are assembled to form a two-dimensional (width and height) image, as illustrated here with a linear array transducer. With linear array devices, multiple "ice-pick" views are lined up side-by-side, like the teeth of a comb.

An array transducer contains multiple elements that can be fired singly or in groups. The elements of a linear array are fired beginning at one end and proceeding to the other. The result is a series of "echo lines" that collectively are called a frame. When the frame is complete, it is displayed on a television screen, and the process of accumulating a new set of echo "lines" begins all over.

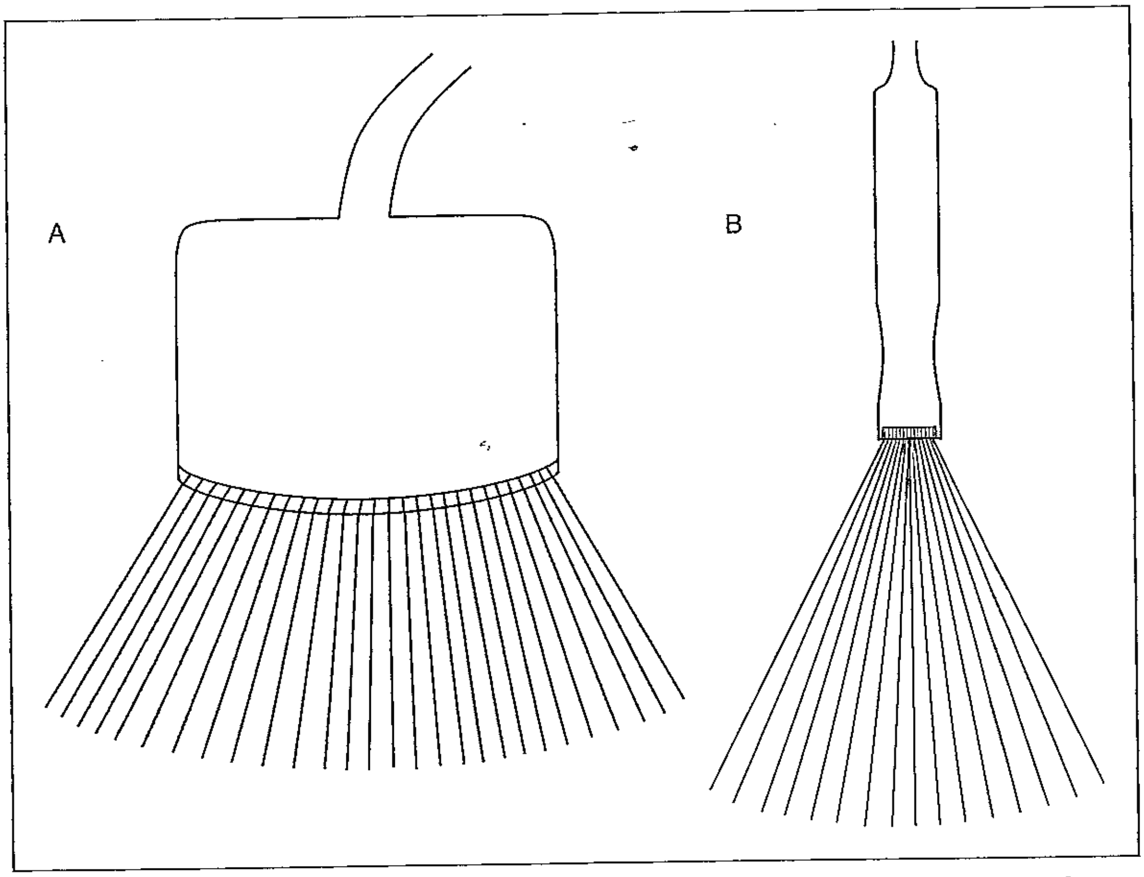


Figure 1-12—Sector Images. Pie-shaped images resembling a *sector* of a circle are used commonly in medical sonography. These images are produced in two ways: **(A)** A curved array may be used. Such arrays operate in the same way as linear arrays, but the transducer elements are oriented on a curved surface. **(B)** A second technique for creating a sector image involves a mini-linear array in which the elements are fired in a certain way to generate a series of radially oriented ultrasound beams (see subsequent figures). With either system, multiple one-dimensional lines of echo information are obtained and stored for subsequent assembly as a two-dimensional image.

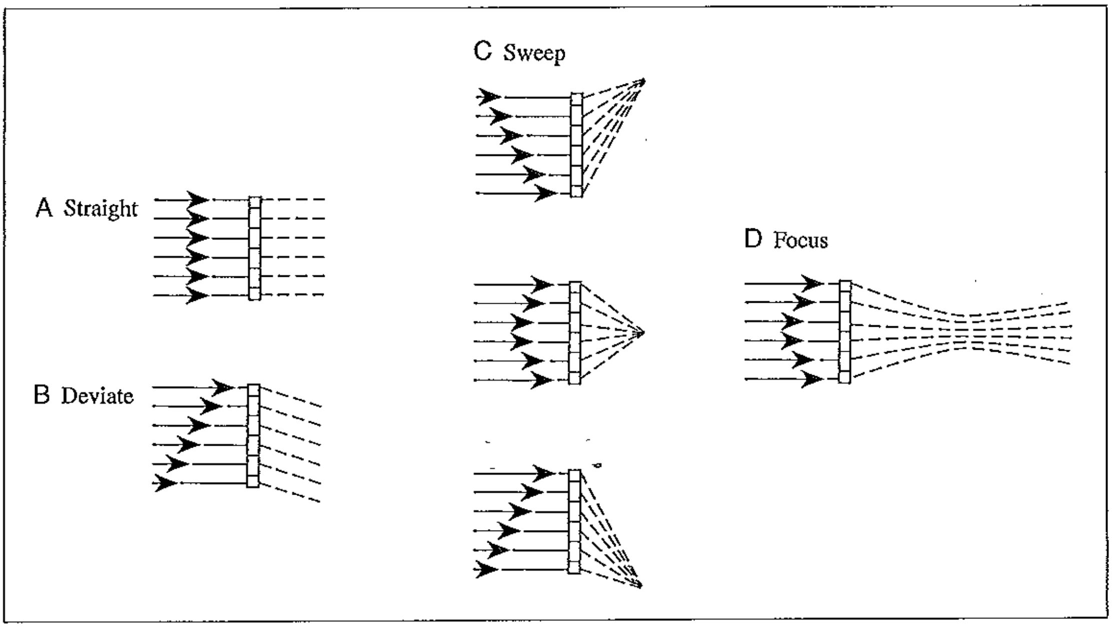


Figure 1-13—Electronic Beam Formation, Steering, and Focusing. Many ultrasound instruments utilize a "phased array" ultrasound crystal to create, steer, and focus the ultrasound beam. Each crystal element produces a "wavelet," or little ultrasound wave. The wavelets merge at a short distance from the crystal to form a unified "*wavefront.*" (**A**) If all the array elements are fired simultaneously, the resultant wavefront moves straight outward from the crystal. (**B**) If each succeeding crystal element is fired a little later than the next, then each wavelet is out of phase with the adjacent wavelet (i.e., the peaks and valleys do not line up). This lack of synchronization causes the wavefront to deviate from a straight path. By changing the element firing delay, the wavefront may be made to deviate a little or a lot. (**C**) For sector image formation (as shown in Figure 1–12*B*), the firing delay is altered sequentially, causing the ultrasound beam to sweep through a sector of a circle. As the beam sweeps through its arc, multiple pulses of ultrasound are sent out along different lines of sight and ultrasound echoes are acquired for each of these lines. These echo lines are stored for display as a two-dimensional image. (**D**) The firing delay mechanism also may be used to focus the beam. This feature is operator controlled, permitting the focal zone to be placed in areas of interest. An analogous method is used to "return focus," or to enhance crystal sensitivity for echoes that arise from specified depths within the body.

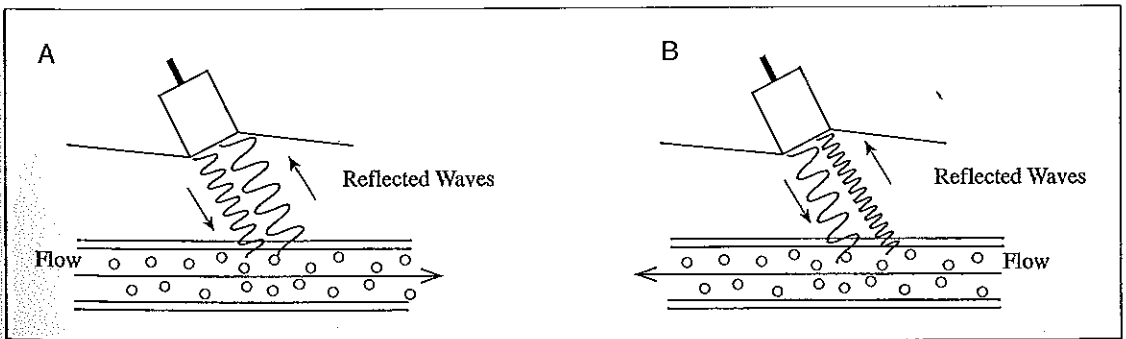


Figure 1-14—Doppler Ultrasound Principles. The Doppler effect refers to the *change* in frequency that occurs when ultrasound is reflected from moving objects. (**A**) If the objects (e.g., red blood cells) are moving *away* (relatively) from the transducer, then the reflected (returning) waves have a *longer* wavelength and a *lower* frequency than the incident (outgoing) waves. (**B**) If the reflecting objects are moving *toward* the transducer, then the reflected waves have a *shorter* wavelength and a *higher* frequency than the incident waves. The frequency difference between the incident and reflected wave is called the *Doppler shift.* The size of the Doppler shift is directly proportionate to the *velocity* of the reflectors. *In essence, Doppler ultrasound provides two pieces of information: (1) the direction of the Doppler shift (increase or decrease) indicates the direction of blood flow; and (2) the size of the shift indicates the velocity of blood flow.*

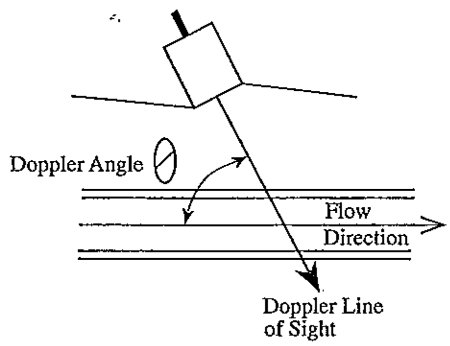
A

$$\text{Doppler Frequency Shift} = \text{Incident* Ultrasound Frequency} - \text{Reflected Ultrasound Frequency}$$

B

$$\text{Doppler Frequency Shift} = \left(\frac{2 \times \text{Incident* Frequency} \times \text{Blood Velocity}}{\text{Constant}} \right) \text{Cosine } \theta$$

C



* The incident frequency refers to the ultrasound beam sent into the body.

Figure 1-15—The Doppler Formula. (A) The Doppler frequency shift is the frequency difference between the incident (outgoing) ultrasound beam and the returning echoes. (B) The Doppler frequency shift may be predicted using the *Doppler formula,* which is illustrated here. Since the incident frequency and the constant are known, the Doppler frequency shift is proportionate to two unknown variables: the velocity of the reflector (blood), and the Doppler angle. (C) *The Doppler angle* is illustrated with the Greek letter theta (θ). The Doppler angle can be determined with modern ultrasound instruments, permitting the Doppler equation to be solved (electronically) for the only remaining variable, the reflector velocity (the velocity of blood flow).

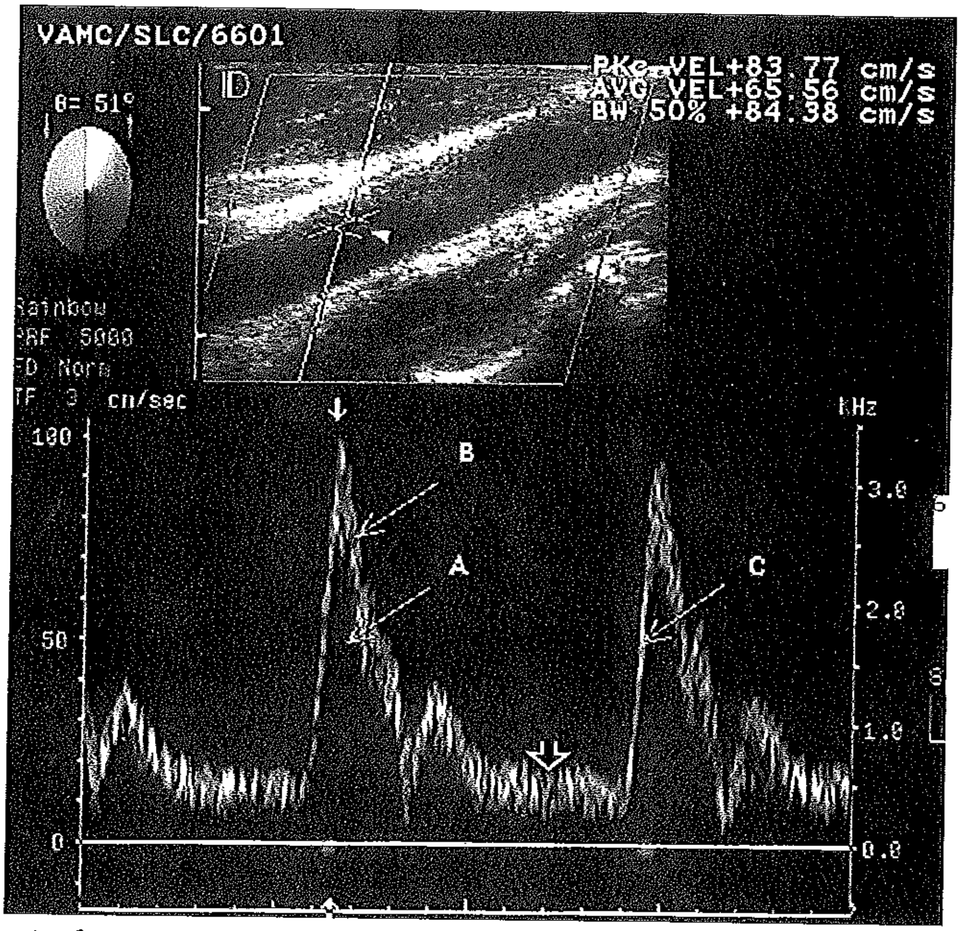


Figure 1-16—Duplex Sonography. The term *duplex sonography* refers to the simultaneous display of an ultrasound image and Doppler information. The display screen typically shows the following information:

B-mode image—A two-dimensional, B-mode image of the area of interest is shown at the top of the display.

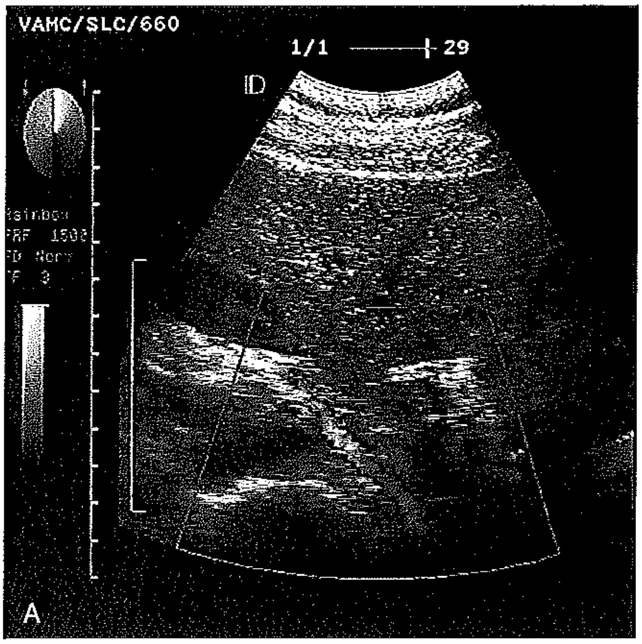
Sample volume—The region where Doppler information is obtained is called the sample volume. This is shown by parallel lines on the B-mode image (arrowhead).

Doppler angle—the Doppler angle (see Figure 1–15C) is visible on the B-mode image and also is displayed numerically (θ = 51 degrees, upper left corner of display).

Flow direction—Flow away from the transducer is shown above the spectrum baseline, and flow toward the transducer is shown below the baseline. (In this case, all flow is away from the transducer; i.e., above the baseline.)

Velocity distribution—the z axis, or the brightness of the spectral display elements, and the width of the spectrum, correlate with the distribution of velocities (frequency shifts) in the sample volume. To better understand the z axis, imagine that the spectrum display is made up of tiny pixels, or picture elements, with each corresponding to a specific moment in time and a specific velocity (frequency shift). At the moment in time indicated by arrow A, the picture elements corresponding to 50 cm/sec are black, indicating that no blood cells are moving at 50 cm/sec. At the same moment, pixels corresponding to 75 cm/sec (arrow B) are gray, indicating that a moderate number of blood cells are moving at that velocity. Picture elements corresponding to arrow C are white, indicating that a large proportion of blood cells are moving at about 50 cm/sec at that moment in time. Note also that the spectrum is very narrow (just a thin line) at C, indicating that at that moment in time most of the blood is moving at the same velocity. In contrast, the spectrum is much thicker (broader) at B, indicating that the range of velocities is much wider at moment B. Note also that a wide range of velocities is present throughout diastole, as indicated by a broad diastolic spectrum (open arrow).

Moment-by-moment velocity data—The vertical arrow at the bottom of the spectrum display represents a moment in time. The numbers at the top right of the display show the maximum velocity (PKc VEL), the average velocity (AVG VEL), and velocity range (BW 50%) within the sample volume at the moment indicated by the vertical arrows.



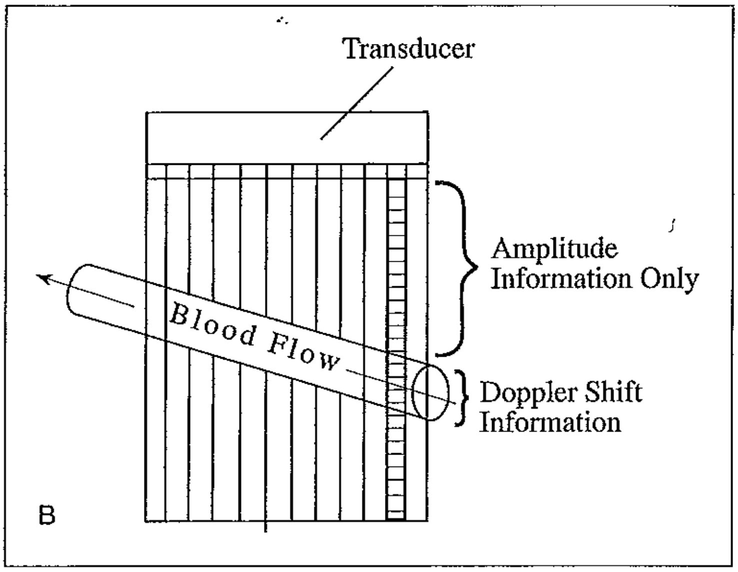


Figure 1-17—Principles of color Doppler sonography. (A) Color Doppler sonography refers to the representation of blood flow in color on a two-dimensional ultrasound image, as shown here. See this figure in the Color Plates. (B) Color Doppler images are produced as follows: Stationary reflectors generate echo amplitude information but do not generate a Doppler shift. Stationary reflectors, therefore, are shown in shades of gray. Flowing blood generates a Doppler shift *in addition* to amplitude information, and all Doppler-shifted echoes are shown in color. The direction of flow is shown by the color hue. For instance, red may represent flow in one direction (relative to the transducer), and blue may represent flow in the opposite direction. The velocity of blood flow is represented by the shade of color. For instance, lower velocities might be dark red, and higher velocities pink. Note the color wheel in the left upper corner of part **A**. This wheel indicates the coding for flow direction and velocity. (See Color Figure 1–17A following page 108.)

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Chapter 2

Image Optimization, Ultrasound Artifacts, and Safety Considerations

William J. Zwiebel

This chapter provides instruction on how to adjust an ultrasound instrument to produce the clearest and most informative ultrasound images. It also provides information on commonly encountered ultrasound artifacts that can be beneficial or problematic diagnostically. Finally, this chapter considers the bioeffects of ultrasound and ultrasound safety. Many of the terms used in this chapter are defined in the preceding chapter, and the reader should refer to these definitions as needed.

IMAGE OPTIMIZATION

Computerized automation has made it relatively easy to adjust ultrasound instruments; nonetheless, sonographers must be familiar with basic principles of ultrasound physics and instrumentation to optimize the function of diagnostic ultrasound devices. Many instruments now are equipped with preset power, gain, and echo-processing parameters that are tailored to specific applications, such as obstetric or abdominal diagnosis. In spite of these conveniences, it remains necessary to "fine tune" the image repeatedly in the course of most sonographic examinations.[1-3]

Six steps are suggested for image optimization:

1. Select a *scanhead* (see definition in preceding chapter) with an ultrasound frequency appropriate to the depth of operation.
2. Choose the appropriate *preprogrammed instrument setup* (e.g., abdomen, pelvis, superficial structures).
3. Adjust the *field of view* to encompass the area of interest.
4. Adjust the *focal depth*.
5. Adjust the *gain and power* settings to provide a uniform level of echoes throughout the image.
6. Select *pre- and postprocessing* settings as needed, to enhance specific image features.

For convenient reference, these six steps are listed in Table 2–1. Each of the six steps will be considered in turn.

Scanhead Selection

Each scanhead contains a piezoelectric crystal that produces a range of ultrasound frequencies, but this range is limited. Different scanheads are required, therefore, for different ultrasound tasks. For example, a 7.5- or 10-MHZ scanhead is ideal for examining the thyroid gland, but these scanheads would be useless for examining the liver or kidneys. At most, a scanhead may be designed to function at three frequencies (e.g., 2.5, 3.5, and 5 MHz); therefore, the need to switch from one scanhead to another is inevitable. Image resolution with higher ultrasound frequencies, such as 5 to 10 MHz, generally is superior to resolution with lower frequencies (2 to 4 MHz), but high-frequency ultrasound cannot penetrate deeply into the body since the ultrasound beam is attenuated (weakened) more

Table 2-1. Ultrasound Instrument Adjustment Protocol

1. Scanhead
   Select a scanhead with a frequency appropriate to the depth of examination.
2. Programmed Setup
   Select the appropriate preprogrammed instrument setup.
3. Field of View
   Adjust the field of view such that the display screen is filled with useful information.
4. Focal Depth
   Place the focal region in the area of maximum interest.
5. Time-compensated Gain (TCG) and Output Power
   First adjust the TCG, and then adjust the output power to produce a uniformly bright image and optimal echo strength.
6. Pre- and Postprocessing
   Select the pre- and postprocessing settings to enhance areas of interest.

readily than a low-frequency beam. In choosing a scanhead, therefore, a compromise must always be struck between resolution and attenuation. Generally, 7.5- to 10-MHz scanheads are used for very superficial structures, such as the thyroid gland or breast; 5.0-MHz scanheads are used for intermediate depths; and 3.0-MHz scanheads are used for deeper structures, such as the abdominal contents. In some cases, even lower frequencies (2 to 2.5 MHz) are useful for abdominal examination in large patients, particularly when Doppler information is desired.

The shape of the ultrasound image also must be considered in scanhead selection. Sector scanheads with a small "footprint," or area of skin contact, are desirable for working in close confines, such as between the ribs. In contrast, a broad nearfield, as provided by a linear or curved array, is advantageous for examining superficial structures such as the thyroid.

The final factor to be considered in scanhead selection is the inherent resolution of a particular scanhead design. In general, linear array scanheads offer better resolution than curved arrays, and curved arrays offer better resolution than sector scanheads. When high resolution is desired (e.g., obstetric imaging), the sonographer should take advantage of the superior resolution of linear arrays wherever possible.

Programmed Settings

The instrument settings that are optimal for different ultrasound examinations vary widely. For example, fetal echocardiography requires the following: (1) a high pulse repetition frequency and frame rate, needed for visualizing rapidly moving structures; (2) a high-contrast, "edge-enhanced" gray scale image to visualize minute cardiovascular structures; and (3) a color Doppler velocity range appropriate for intracardiac flow. These instrument settings are vastly different from those that optimize abdominal images; namely: (1) a relatively low pulse repetition frequency and frame rate (abdominal viscera do not move fast); and (2) a broad gray scale image that enhances "tissue texture" resolution.

Because it can be quite difficult to arrive at the right combination of instrument settings for a given examination, ultrasound instruments are preprogrammed for specific applications, such as abdominal sonography, peripheral vascular examination, and obstetric sonography. The sonographer can pick and choose among these applications as needed. It is important to use the proper preset adjustment package for the examination at hand, or else image quality may suffer greatly. If the preset programs do not fit the needs of a given ultrasound department, then they can be modified as needed, or entirely new programs can be created for specific applications. The manufacturer can provide assistance and recommendation for such program modifications.

Field of View

The image size should always be adjusted so that the display screen is filled with useful information, as illustrated in Figure 2–1. Do not be a "postage stamper," who scrunches a tiny image into the corner of the display! Conversely, do not magnify the image so greatly that orientation is impossible.

Focus

The term "focus" is used broadly, since ultrasound "focusing" involves electronic techniques that may be quite different from optical focusing. Nevertheless, two important points must be made concerning ultrasound focusing: First, maximum resolution occurs in the focal zone, and resolution may decrease significantly at a distance from the focal zone—therefore, the focal zone should always correspond to the area of maximum diagnostic interest (Fig. 2–2). Second, if a broad field of resolution is desirable, multiple focal areas should be used rather than a single focal zone; the trade-off, however, is a reduction of frame rate.

Time-Compensated Gain

As an ultrasound beam travels through the body, two processes diminish the strength or "intensity" of the beam: (1) the beam is attenuated, meaning that mechanical energy (vibration) is converted to heat; and (2) the beam becomes dispersed or spread out as it is reflected and refracted at acoustic interfaces. Because of attenuation and dispersion, distant ultrasound echoes are *much* weaker than near echoes; therefore, distant echoes must be amplified much more than near echoes to produce an image that is uniformly bright from the nearfield to the farfield (from top to bottom) (Fig. 2–3). The term "time-compensated gain" describes the process of amplifying distant echoes more than near echoes. Stated literally, the *gain* (amplification) is increased to *compensate* for the effects of transit *time.* Since distance equals time in ultrasound parlance, we also could say that gain is increased to compensate for distance from the scanhead.

Ultrasound attenuation varies markedly as the scanhead is moved from one position to another in the course of an ultrasound examination, and

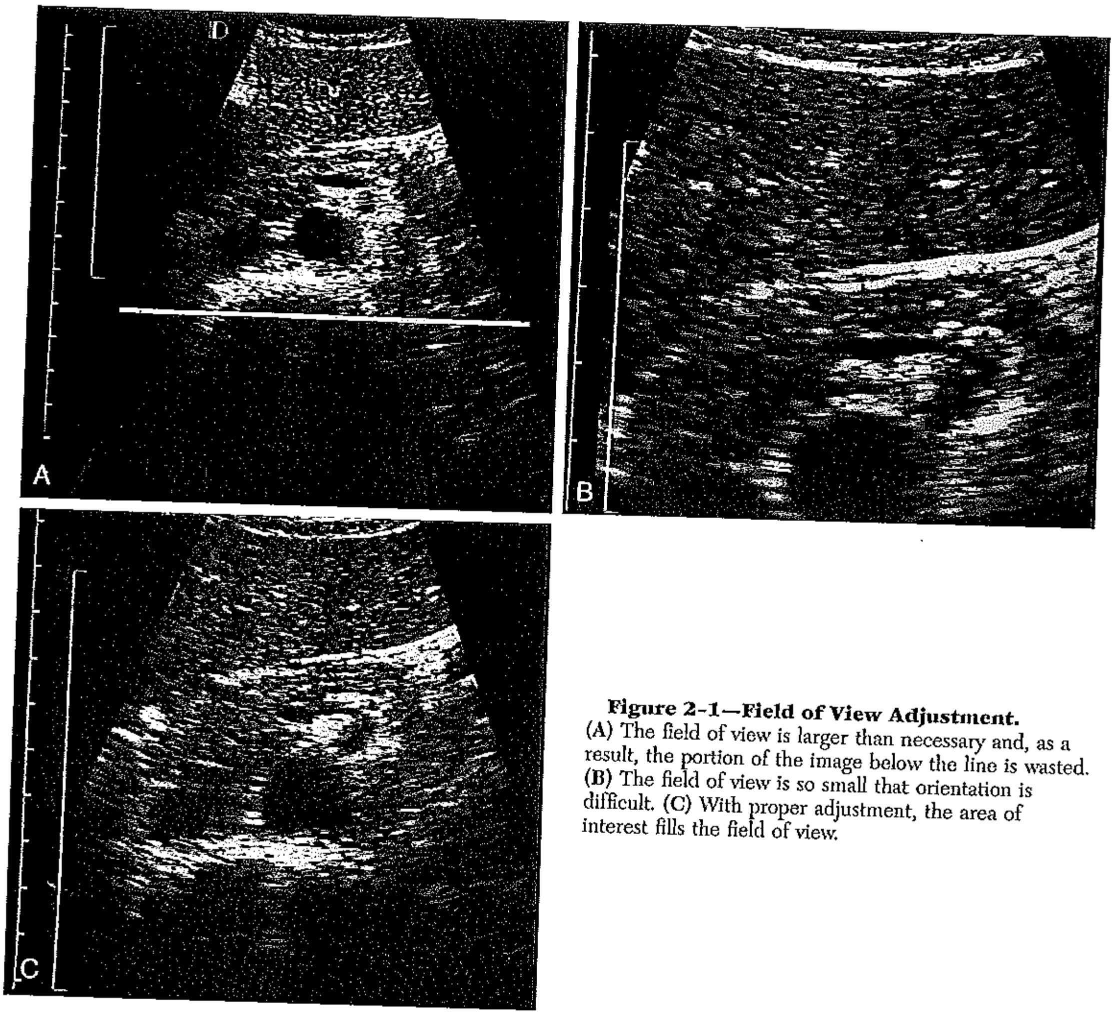

Figure 2-1—Field of View Adjustment. (A) The field of view is larger than necessary and, as a result, the portion of the image below the line is wasted. (B) The field of view is so small that orientation is difficult. (C) With proper adjustment, the area of interest fills the field of view.

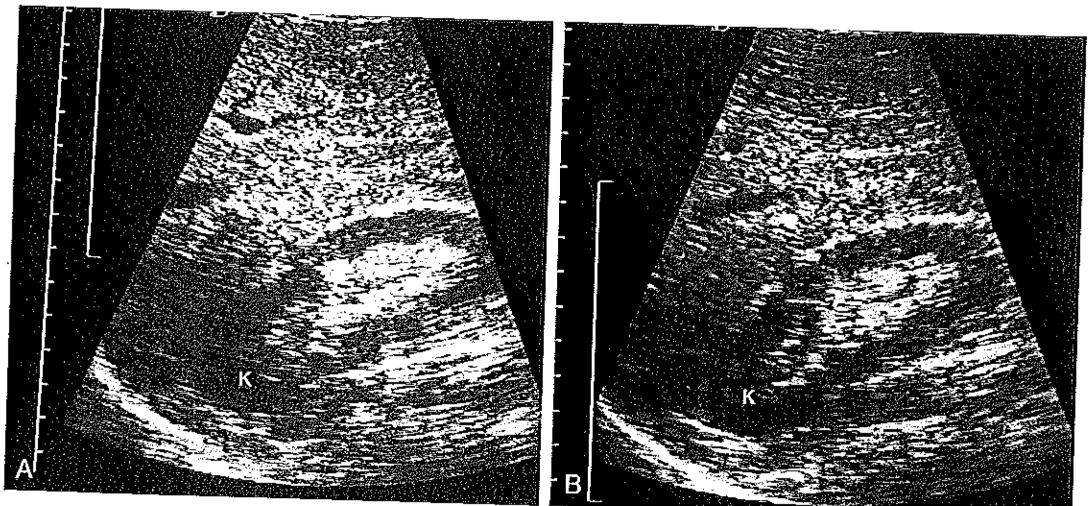

Figure 2-2—Effects of Focal Zone Adjustment. (A) The focal zone (bracket at left of image) is too high, and as a result the superior pole of the kidney (K) is indistinct. (B) With the focal zone properly positioned, the upper pole of the kidney (K) is clearly seen.

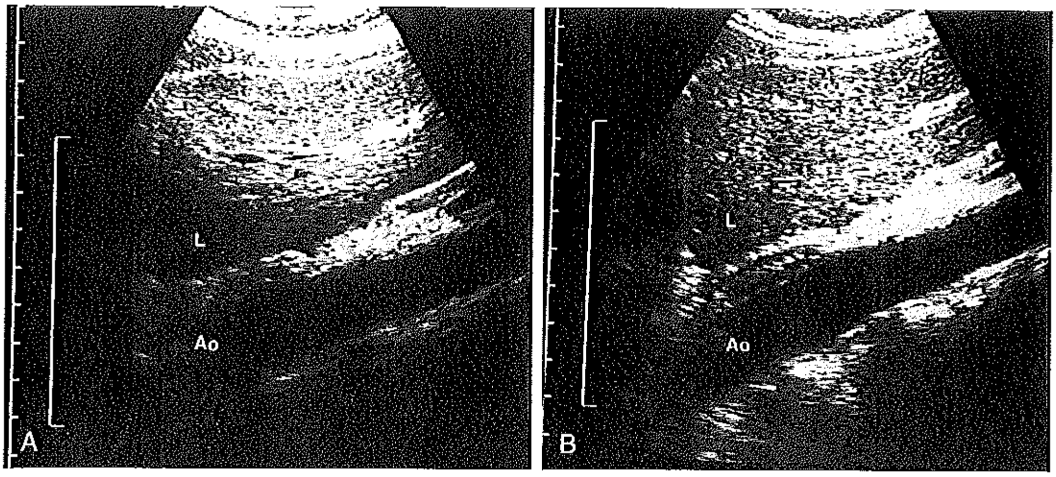

Figure 2-3—Time-Compensated Gain. (A) In this image, the TCG is improperly set, resulting in poor visualization of the posterior portion of the liver (L) and the aorta (Ao). (B) With proper TCG adjustment, the liver (L) and the aorta (Ao) are seen clearly.

time-compensated gain (TCG*) must therefore be adjusted *repeatedly* to maintain uniform image intensity. Proper TCG adjustment is particularly important in the pelvis to insure clear visualization of the female reproductive organs through the distended bladder.

Power

The strength, or power, of the ultrasound beam sent into the patient may be adjusted by the sonographer. The power of the beam is measured in absolute terms as watts per square centimeter. Ultrasound power also is measured with a relative term called the decibel (dB), and this term is particularly important since it typically is used for instrument power settings. The decibel is a logarithmic expression that *compares* the power of two ultrasound beams. If two beams differ in intensity by 1 dB, then the power difference is tenfold; that is, one beam is ten times more powerful than the other. Thus, 2 dB equates to a 100-fold difference in power, and 3 dB equates to a 1000-fold difference, and so on. Output power controls on ultrasound instruments frequently are calibrated in dB, *relative to the maximum or minimum output of the instrument and scanhead in use.* The term "relative" is crucial; the power output settings of one instrument do not equate to those of another instrument, and the output of one scanhead does not equate to that of another scanhead. The *actual* power output may vary greatly from one instrument or scanhead to another. Many instruments are programmed to prohibit output power levels above certain limits, per the requirements of the United States Food and Drug Administration.

Image quality is affected significantly by output power, but it is not necessarily true that more power equates to superior image quality. Output power must be tailored to match the scanhead and the echo processing components. It is best, therefore, to begin with the power setting that is preprogrammed for a specific diagnostic application. If visualization of deep structures is limited even though TGC is maximized, then it may be advisable to increase the output power (Fig. 2–4). If an increase in output power merely fills the image with noise, then no benefit is derived, and it may be preferable to switch to a lower frequency scanhead to reduce ultrasound beam attenuation.

Preprocessing and Postprocessing

The echo information that returns from the patient is processed in two ways before the image is assembled on the display screen. First, the "raw" echo signal from the transducer is "preprocessed," prior to storage of the echo information; second, the stored echo information is "postprocessed," prior to display on the image screen.

Preprocessing principally determines which echoes are saved and which are discarded, on the basis of echo strength. For example, with one preprocessing program, high-intensity echoes are stored and low-intensity echoes are discarded. With another program, only the low-intensity echoes are saved, and with yet another program, echoes are retained over a wide range of intensities.

*The abbreviations TCG, for time-compensated gain, and TGC, for time gain compensation, are equivalent and may be used interchangeably.

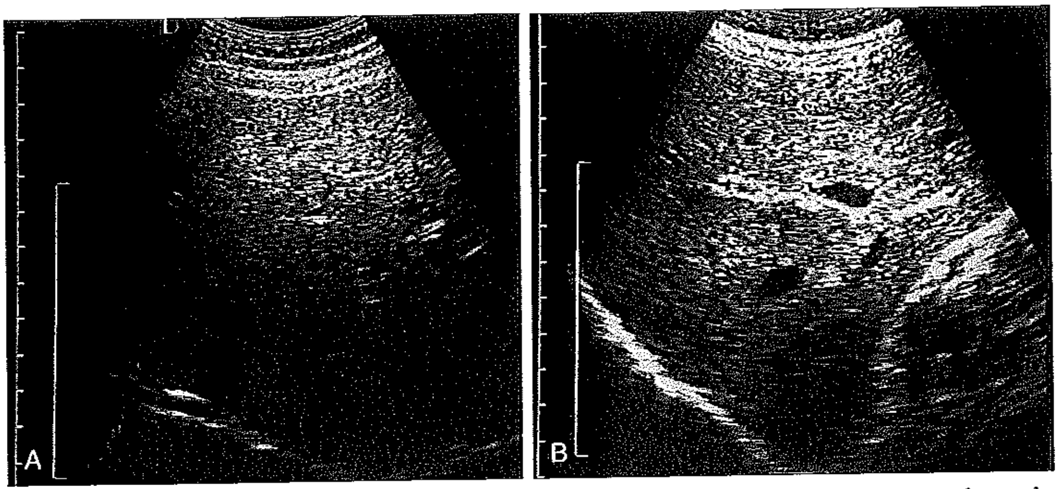

Figure 2-4—Effects of Output Power Adjustment. (A) The TGC has been maximized in this example, yet the posterior aspect of the liver and the diaphragm are poorly seen because output power is low. (B) With increased output power, and readjustment of the TGC, the posterior aspect of the liver is seen clearly.

Postprocessing concerns the selection of stored echoes for display on the image monitor. This selection process is based on echo amplitude or gray scale level. To understand postprocessing, consider that a television monitor may display 32 shades of gray but that a much broader range of echo intensities exists in storage. A decision must be made as to which echoes are shown on the display and which are not. One option might be to show the entire range of echoes but to compress them into 32 gray shades. The resultant image would have wide latitude, but subtle gray scale differences would be lost. Another option might be to show primarily the low-level echoes. This approach would enhance subtle tissue "texture" differences, but the image might be "noisy" and indistinct. Finally, high-intensity echoes might be displayed while the low-intensity echoes are ignored. The latter approach would emphasize structural borders, such as the edges of blood vessels, but gray scale "texture" might be lost. The potential effects of postprocessing are illustrated in Figure 2–5.

From a practical perspective, it is best to begin pre- and postprocessing adjustments with the programmed settings provided by the manufacturer. Preprocessing generally is left at the programmed setting, and only postprocessing is adjusted to enhance specific components of the image. Postprocessing is particularly useful since frozen images can be postprocessed "after the fact" to emphasize areas of interest.

IMPORTANT ULTRASOUND ARTIFACTS

As noted earlier, several commonly occurring ultrasound artifacts affect ultrasound interpretation, either positively or negatively. There is a vast array of ultrasound artifacts, but we will consider only the most important ones; namely, enhanced through-transmission, acoustic shadowing, lateral edge shadows, reverberation artifacts, slice thickness artifacts, and "side lobe" artifacts.[1-11]

Enhanced Through-Transmission

Earlier in this chapter, we considered the need to adjust the time-compensated gain (TCG) so that weaker echoes from deeper structures are amplified more than stronger echoes from near structures. The TCG adjustment assumes that ultrasound attenuation is relatively uniform throughout the tissues imaged, but attenuation actually varies considerably from one point to another in most clinical applications. If the ultrasound beam passes through an area of unexpectedly low attenuation, deeper structures receive stronger ultrasound pulses and produce stronger echoes than the gain settings anticipate. As a consequence, echoes from the deeper structures are overamplified and are much brighter than other echoes at a similar depth (Fig. 2–6). The resultant artifact is called "enhanced through-transmission," since ultrasound transmission is enhanced in the low-attenuation area.[1-6]

Enhanced through-transmission is a crucial diagnostic feature of fluid-filled structures such as cysts and abscesses. A mass should be regarded as fluid-filled only if enhanced through-transmission is evident. A low-attenuation mass that does *not* show enhancement may be a homogeneous

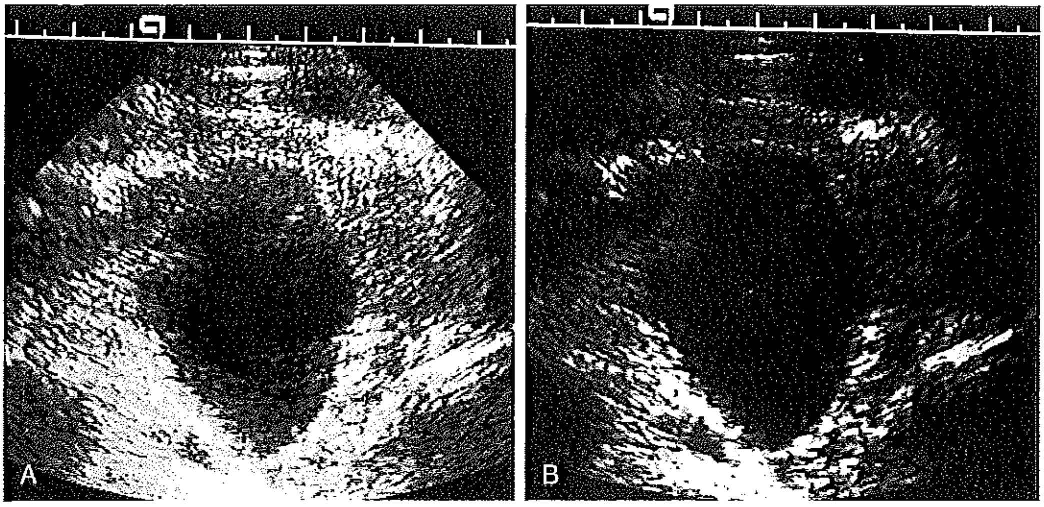

Figure 2-5—Effects of Postprocessing. (Subject: pelvic cyst.) (A) Postprocessing is adjusted to display a wide range of echo intensities that makes the image somewhat cluttered. (B) Postprocessing is adjusted to display only high-level echoes, which produces a sharper, "edge-enhanced" image, but low-level echoes are lost.

solid mass (e.g., lymphoma), rather than a fluid-filled structure.

Acoustic Shadowing

The converse of enhanced through-transmission is acoustic shadowing.[1–5, 7] In this case, a highly reflective or highly attenuating area blocks transmission of the ultrasound beam, leading to weak or absent distal echoes (Fig. 2–7). The result is a hypoechoic or echo-free area, called an acoustic shadow, distal to the highly reflective or highly attenuating structure. Acoustic shadows can hinder visualization of important structures, but they also can be of great diagnostic value, as is the case with gallstones. They are an essential diagnostic feature of gallstones, and an object in the biliary tree should not be regarded as a gallstone if it does not produce an acoustic shadow.

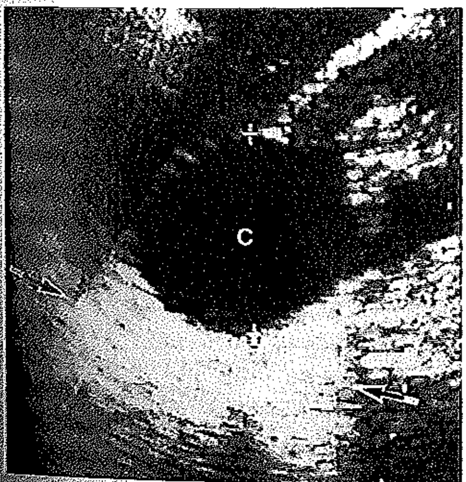

Figure 2-6—Enhanced Through-Transmission. A band of high-intensity echoes (arrows) is present distal to a renal cyst (C). The intensity of these echoes is greater than that of other echoes at similar depths because the ultrasound beam is attenuated very little as it traverses the homogeneous cyst fluid.

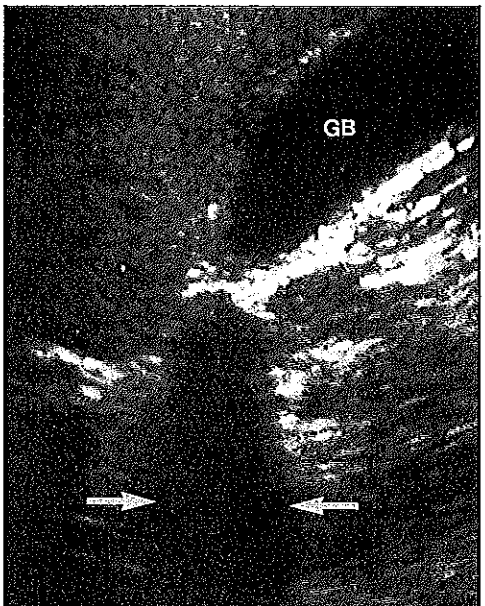

Figure 2-7—Acoustic Shadow. A distinct shadow (arrows) is seen distal to a gallstone impacted in the gallbladder (GB) neck. The shadow is caused by virtually complete deflection of the ultrasound beam by the gallstone.

Lateral Edge Shadows

Acoustic shadows may occur distal to the walls of well-encapsulated lesions, as a result of reflection and refraction of the ultrasound beam in the lesion wall (Fig. 2–8). These shadows, which are called lateral edge shadows,[1–5, 7] are a useful diagnostic feature of cysts but are seen only when high-frequency, high-resolution instruments are used. Lateral edge shadows are not specific for cysts, for such shadows also occur with well-encapsulated solid lesions.

Reverberation Artifacts

Reverberations are common and annoying artifacts that occur in the form of parallel lines oriented perpendicular (more or less) to the ultrasound beam[1–5] (Fig. 2–9). They are generated when an ultrasound beam bounces back and forth between strong specular reflectors, or between a specular reflector and the transducer. The net result is an increase in the time required for the ultrasound beam to get back to the transducer (remember, time is distance in ultrasound). Several features serve to identify reverberation artifacts: (1) they usually are multiple; (2) they are parallel and repeat at regular intervals; (3) one or more strong specular reflectors often can be identified as the source of the artifact; and (4) succeeding reverberations become weaker, since the beam is attenuated as it bounces back and forth between the reflectors.

Comet Tail Artifacts

Reverberations also may occur *within* small, very dense objects, such as small metal fragments or cholesterol crystals, as illustrated in Figure 2–10. A characteristic "comet tail" artifact results from these internal reverberations. This artifact consists of short parallel lines that taper and "fade out" distal to the object.[1–5, 8, 9]

Slice Thickness Artifacts

An ultrasound image is shown on a television monitor as if it has only two dimensions, width and height. The ultrasound slice actually has three dimensions, width, height, and thickness, but slice thickness is not shown. As illustrated in Figure 2–11, the compression of three dimensions into two may superimpose objects and cause diagnostic ambiguity.[1–5, 10] Slice thickness artifacts include cysts that appear echogenic and spurious thickening of the gallbladder wall.

Off-Axis Ultrasound Beams

We typically assume that an ultrasound transducer produces a single ultrasound beam that propagates straight outward from the crystal, but

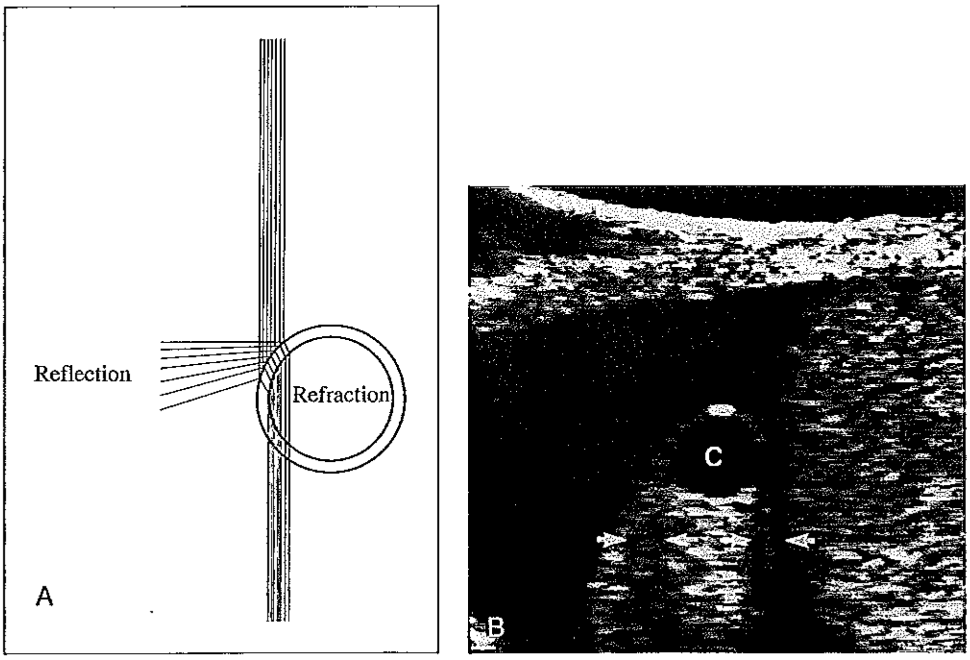

Figure 2–8—Lateral Edge Shadows. (A) Reflection occurs at the surface of a sharply marginated object, and refraction occurs within the wall of the object. These phenomena generate an edge shadow. (B) In this clinical example, lateral edge shadows (arrows) are seen distal to the wall of an epididymal cyst (C).

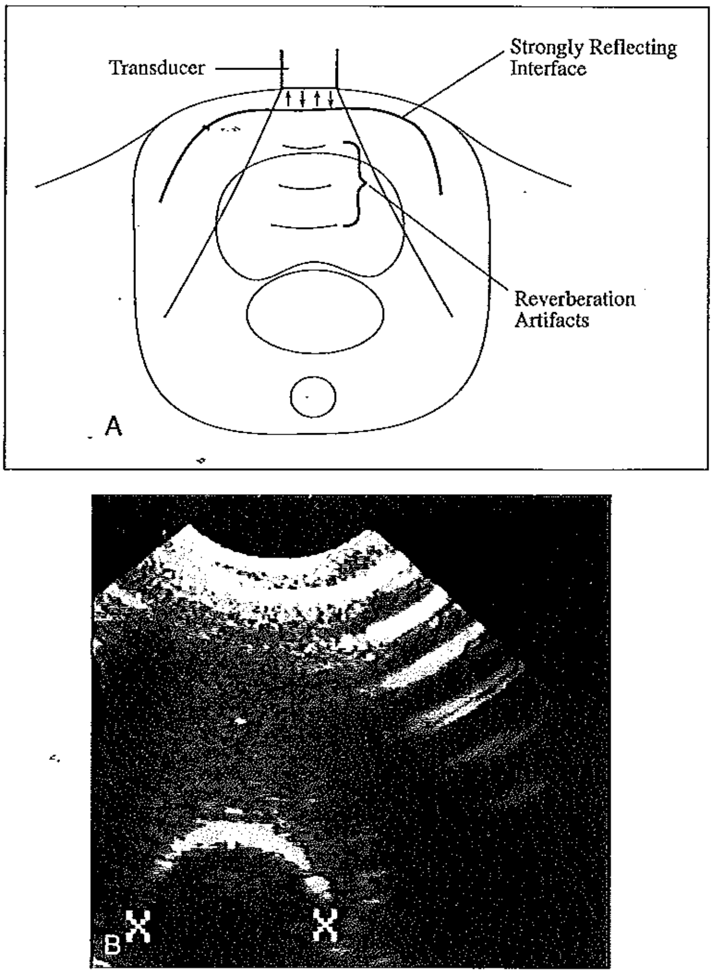

Figure 2-9—Reverberation Artifacts. (A) Reverberation occurs when the ultrasound beam bounces back and forth between strongly reflecting surfaces. In this diagram, the sound beam bounces (arrows) between the transducer face and a strongly reflecting superficial interface. Reverberation artifacts are produced that are superimposed on deeper structures as uniformly spaced parallel lines. Note that the distance between these lines is twice the distance between the reflectors. (B) In this clinical example, prominent reverberation artifacts are seen along the right side of the image. These were produced by reverberation between the transducer and a rib.

this is not really the case. In addition to the main beam, all ultrasound transducers produce off-axis beams that are known by names such as side lobes or grating lobes[1–5, 11] (Fig. 2–12*A*). These off-axis ultrasound beams are significantly weaker than the main beam; nevertheless, they may generate echoes that are mistakenly placed in the midst of the "real" echoes from the main beam. Two different types of artifacts may result from off-axis ultrasound beams. First, and most commonly, the image may be "peppered" with a haze of spurious echoes that reduces detail (Fig. 2–12*B*). Second, spurious structures are superimposed on real structures (Fig. 2–12*C*).

Mirror Image Artifacts

An object in the ultrasound image is represented twice in the mirror image artifact, once in its correct location and a second time in an artifactual "mirror image" location. Mirror image artifacts are most commonly seen adjacent to the diaphragm, where an "extra" diaphragm may appear in the image (Fig. 2–12*C*), or an echoic liver lesion may be shown twice: once in the liver where it belongs and a second time in the lung, where it does not belong (or vice versa).

Mirror image artifacts are caused by two basic fallacies in the way ultrasound machines think.[2,4,5] First, the ultrasound machine assumes that all echoes that return to the transducer arise from structures along the beam axis. This is not true, because off-axis ultrasound beams occur, as illustrated in Figure 2–12. The second fallacy is that the ultrasound beam always travels in a straight line to and from a reflector. This also is not the case, for reflection and refraction may divert the beam. These two fallacies of "ultrasound machine thinking" cause ambiguity in ultrasound distance measurements (range ambiguity). Ambiguous distance measurements lead to misplacement of objects in the ultrasound image. Two additional examples of mirror image artifacts are shown in Figure 2–13.

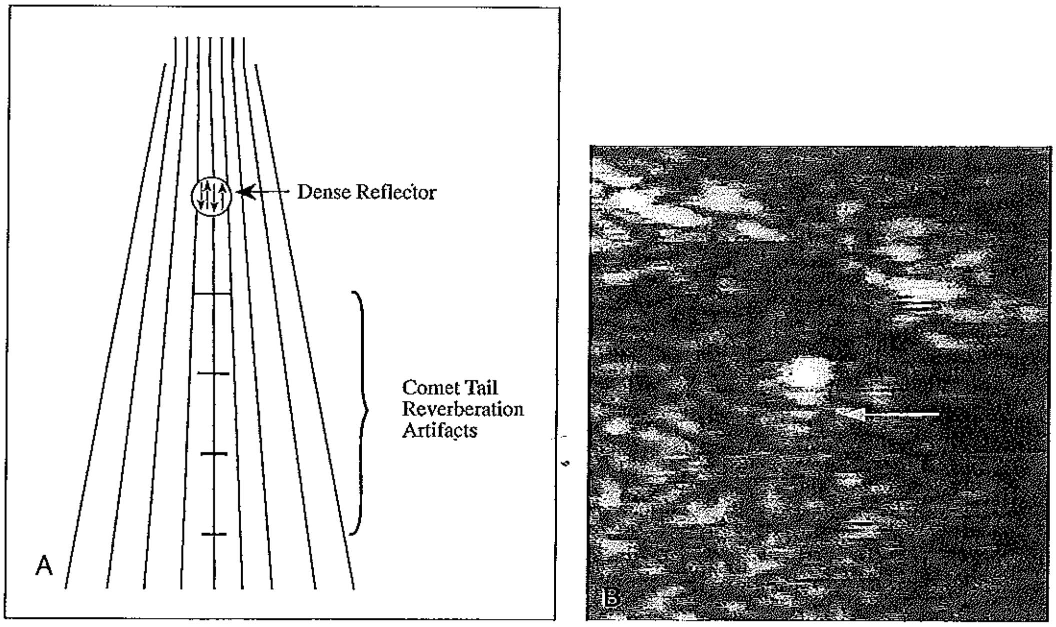

Figure 2-10—Comet Tail Artifacts. (A) Reverberation of the ultrasound beam (arrows) within a dense object produces comet tail artifacts seen distal to the object. (B) In this clinical example, a distinct comet tail (arrow) is seen deep to a dense object (possibly crystalline cholesterol) within the gallbladder.

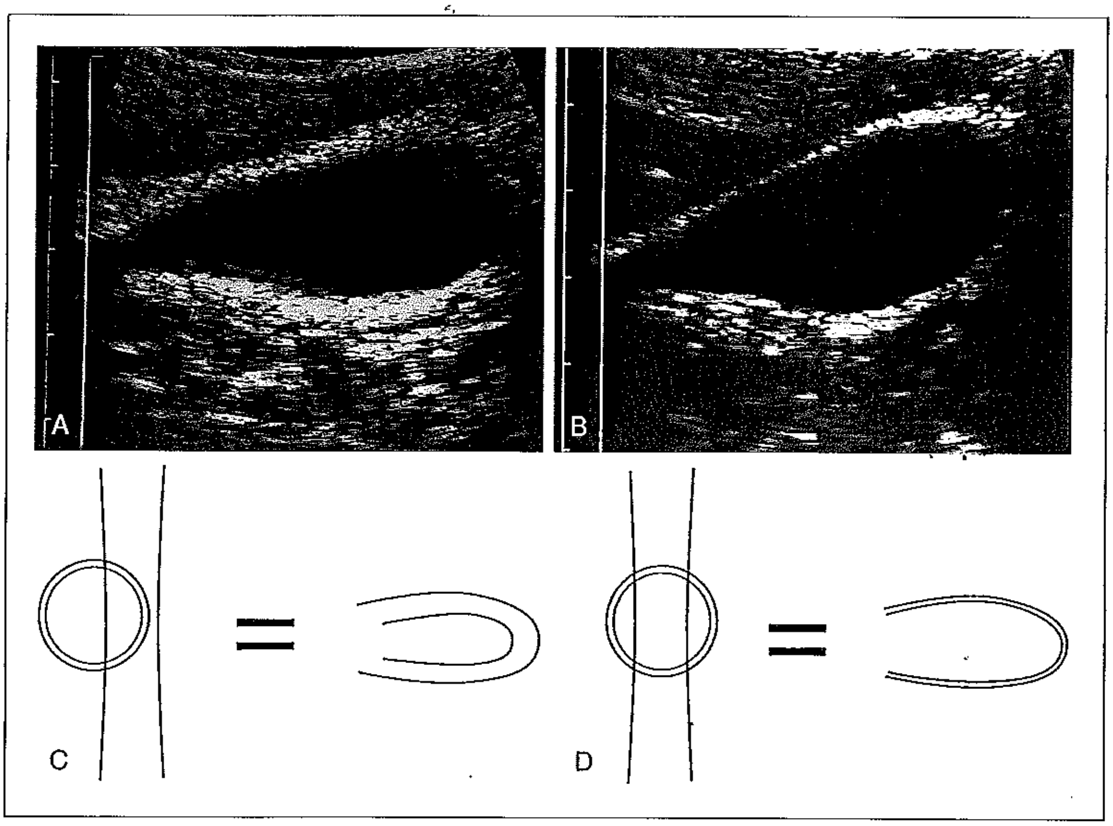

Figure 2-11—Slice Thickness Artifact. (A) The gallbladder wall appears thick in this clinical example, but this thickening is artifactual. (B) With proper position of the scan plane, the gallbladder wall is thin and sharply defined. (C) Artifactual wall thickening in this example occurred when the slice thickness fell partially within the gallbladder lumen and partially outside the lumen. (D) The gallbladder wall was correctly shown when the beam axis passes through the diameter of the gallbladder lumen, minimizing the effect of slice thickness.

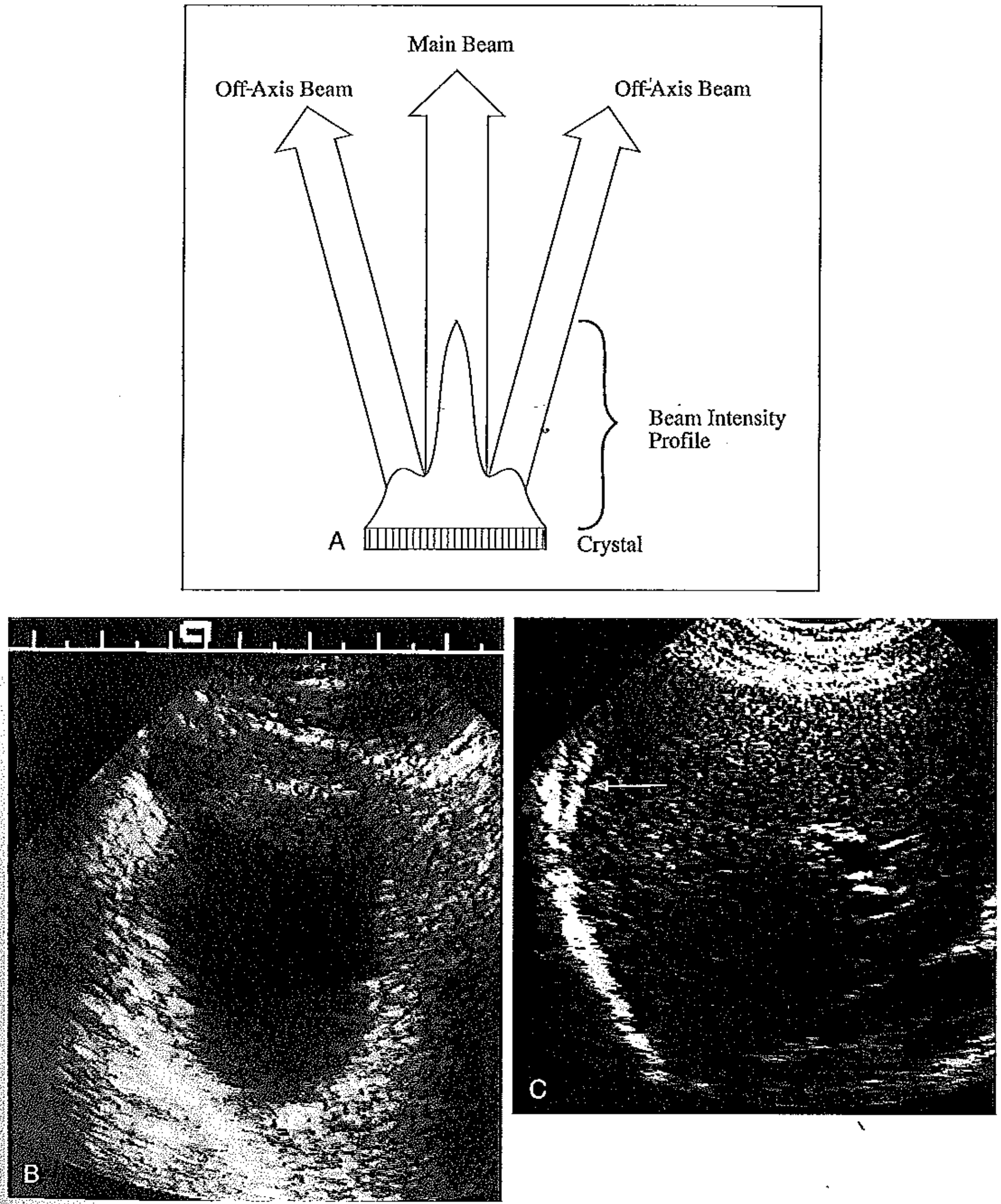

Figure 2-12—Artifacts Due to Off-Axis Ultrasound Beams. (A) Ultrasound transducers produce off-axis, secondary ultrasound beams in addition to the main central beam. (B) The diffuse punctate echoes within this pelvic cyst probably are caused by off-axis ultrasound beams striking reflectors outside the cyst. (C) A mirror image of the diaphragm (arrow) is generated by an off-axis beam that strikes a part of the diaphragm that is not in the image plane.

ULTRASOUND POWER MEASUREMENT

It was noted previously that the power of an ultrasound beam may be quantified with a unit of measure called the decibel. Measurements in decibels are useful for describing the *relative* power output of a device, but the decibel does not have an absolute value, and the *actual* power cannot be described with this term. Decibel calibration is like saying that a vehicle is running at a certain percent of its maximum power; for one vehicle, a given percentage might be 50 horsepower and for another vehicle the same percentage might be 150 horsepower.

The watt is a commonly used *absolute* value

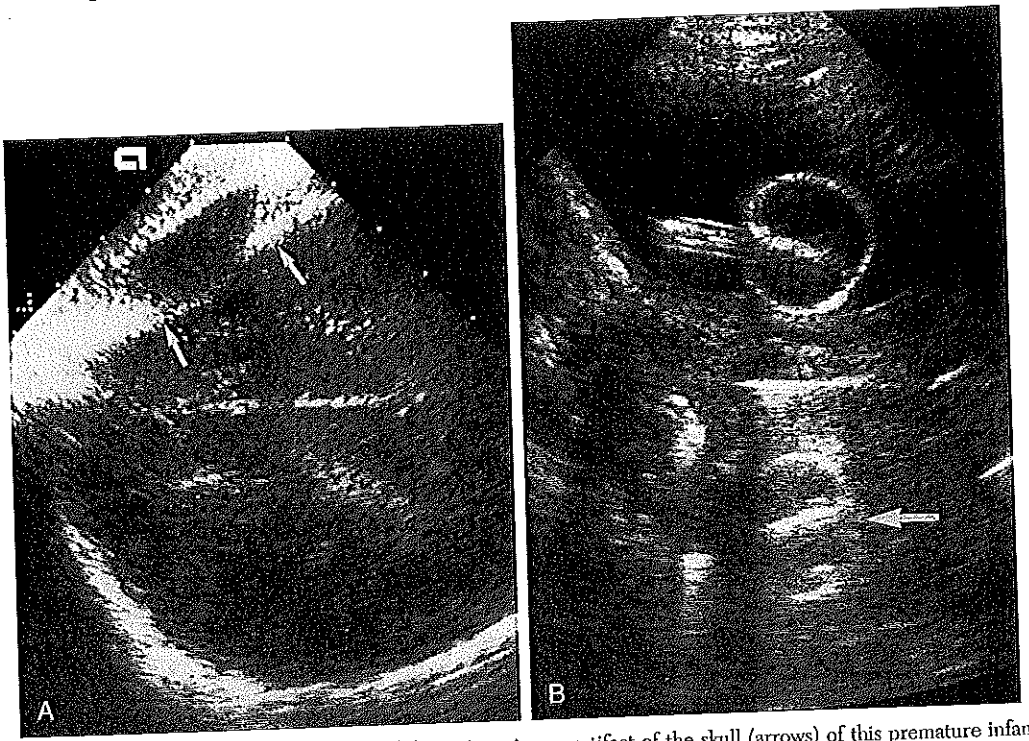

Figure 2-13—Mirror Image Artifacts. (A) A mirror image artifact of the skull (arrows) of this premature infant was mistaken for the margin of an extra-axial fluid collection. No fluid was present on a computed tomogram performed in response to this finding. (B) A mirror image of a Foley catheter (arrow) is seen distal to the real image within the bladder.

for measuring acoustic power, and a number of methods are available for accurately measuring ultrasound power in watts.[3, 12–15] The measurement process is complicated, however, by two factors: (1) power intensity varies markedly from moment to moment in pulsed ultrasound applications, since the ultrasound pulses are interspersed with long "listening" periods; and (2) ultrasound power varies markedly from one point to another within the beam due to the effects of attenuation and focusing. The existence of these factors creates three potential choices for describing pulsed ultrasound power: (1) as a temporal peak or temporal average; (2) as a pulse peak or pulse average; and (3) as a spatial peak or spatial average. The most commonly used measures are spatial peak temporal average (SPTA), reported in milliwatts per square centimeter (mW/cm^2); and spatial peak pulse average (SPPA), reported in watts per square centimeter (W/cm^2).

In the United States, the Food and Drug Administration has authority to regulate diagnostic ultrasound instruments and requires extensive verification of output power for all devices used for medical purposes. A standardized system for displaying output power has not been adopted by manufacturers of ultrasound instruments, but it is likely that such a system will be adopted or required in the future.

It is noteworthy that output power varies markedly among presently available diagnostic ultrasound instruments; furthermore, for a given instrument, output also may vary markedly from one application to another.[12, 14–16] Output power generally is quite low for applications involving linear array transducers. Sector scanners generally require slightly higher power than linear arrays, and spectral Doppler applications require the greatest power output of all. The output power of modern pulsed ultrasound instruments varies from 0.02 mW/cm^2 SPTA for some linear array applications to 4000 mW/cm^2 SPTA for some Doppler applications (note: the high end is 4 $watts/cm^2$!).[12, 15] The power output of diagnostic ultrasound instruments has risen steadily over the history of medical sonography and has taken a particularly great jump in recent years.[12, 15, 16] This recent increase in power has raised additional concern with respect to the safety of ultrasound diagnosis.

ULTRASOUND SAFETY

The safety of ultrasound is of elementary importance and has been the subject of extensive investigation since ultrasound was first used for diagnostic purposes early in the 1950s. Four physical effects of ultrasound are of principal concern: (1) tissue heating resulting from frictional resistance to wave transmission; (2) streaming of suspended particulate matter or cell structures; (3) direct vibratory effects on membranes and other cell structures; and (4) cavitation phenomena.[12–15] The greatest concern has centered on tissue heating and cavitation.

Tissue heating in response to ultrasound is directly proportionate to beam intensity and the duration of exposure. Furthermore, a 1-degree centigrade (C) elevation of tissue temperature is an accepted threshold for tissue damage (i.e., tissue damage is thought not to occur if the temperature rise is under 1 degree C).[13, 14] It is generally accepted that tissue heating occurs in humans during diagnostic ultrasound exposure, but it is not known whether heating ever is sufficient to cause biologic damage.[13]

Cavitation refers to the formation and/or expansion of microbubbles within a liquid medium in response to the rapid pressure oscillations induced by the ultrasound wave.[12–14] During the rarefaction (pressure reduction) phase of an ultrasound beam, microbubbles form and/or expand, creating cavities. These cavities decrease in size or collapse during the compression phase. As the ultrasound waves pass through tissue, massive and rapid alteration of cavity size may occur, potentially causing violent movement or actual streaming of cell contents, as well as other direct physical damage. In the worst case, violent expansion and collapse of cavities causes tremendous shock waves, local heating, and the formation of "sonochemicals," including free radicals.[14]

Cavitation, unlike tissue heating, is not precisely related to beam intensity or exposure time and can occur at quite low ultrasound output levels. No threshold is thought to exist, therefore, for cavitation damage. The generation of cavitation with ultrasound appears to depend on the attainment of very precise conditions, and it is not known whether cavitation occurs in humans during diagnostic ultrasound exposure.[12–14]

Safety Experiments

Concerns about ultrasound safety have been focused on biologic damage that might injure or kill an organism directly, and teratogenic effects that might injure or kill the offspring of an organism. These concerns have generated bioeffects studies that fall into several general categories: (1) mechanistic experiments that address the nature of ultrasound damage (e.g., heat production and cavitation); (2) outcome experiments that address the effects of ultrasound exposure on mammalian and nonmammalian tissues (e.g., insonation of cultured cells, mice, or rats in an attempt to detect adverse effects on the individuals or their offspring); and (3) epidemiologic studies of humans exposed to ultrasound.[12–15] The results of such experiments can be summarized as follows:

1. Potentially damaging physical effects of ultrasound, such as heating and cavitation, have been demonstrated convincingly in vitro. Although these phenomena potentially could occur in humans at current diagnostic ultrasound exposure levels, it has not been demonstrated that these effects in fact do occur, and no potential adverse effects of these phenomena have been revealed by epidemiologic studies.[12–14]
2. Potentially adverse effects of ultrasound at diagnostic levels have been shown in certain tissue culture and laboratory animal studies, but independently verified adverse effects have not been demonstrated at output power levels of 100 mW/cm^2 or less for unfocused beams and 1000 mW/cm^2 or less for focused beams. The American Institute of Ultrasound in Medicine suggests, therefore, that these output power levels are safe for diagnostic purposes. It is postulated, furthermore, that significantly higher output power would be required to produce adverse effects in humans.[13]
3. Epidemiologic studies in humans have not documented any adverse effects of ultrasound exposure, including exposure to fetuses.[13]
4. No adverse effects of diagnostic ultrasound on patients or ultrasound operators have become apparent empirically during more than 40 years of clinical use.[13]

Is Ultrasound Safe?

What about the bottom line: is ultrasound safe? After 40 years of diagnostic ultrasound experience, we can say that we *think* ultrasound is safe, even though we know that ultrasound can cause cellular damage in an in vitro setting at currently used output levels. It is likely that we will never be able to prove conclusively that ultrasound is safe. The increase in output power that has occurred in recent years is noteworthy.[16] Some diagnostic instruments clearly operate above the recognized safe limit of 1000 mW/cm^2. Although it is likely that the upper limit of safe operation exceeds 1000 mW/cm^2,[13] we would do well to be cautious, especially during Doppler examination of fetuses. Output power is apt to be maximum during spectral Doppler examination, and the risk of biologic effects, at least empirically, is at its greatest in the early stages of fetal development. Therefore, in my opinion, fetal ultrasound exposure should be kept as low as possible during the first trimester, and it should particularly be minimized during the period of embryogenesis (the first 9 weeks of pregnancy). In keeping with this philosophy, I feel that fetal Doppler examination should be conducted only when absolutely necessary during the first trimester of pregnancy.

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