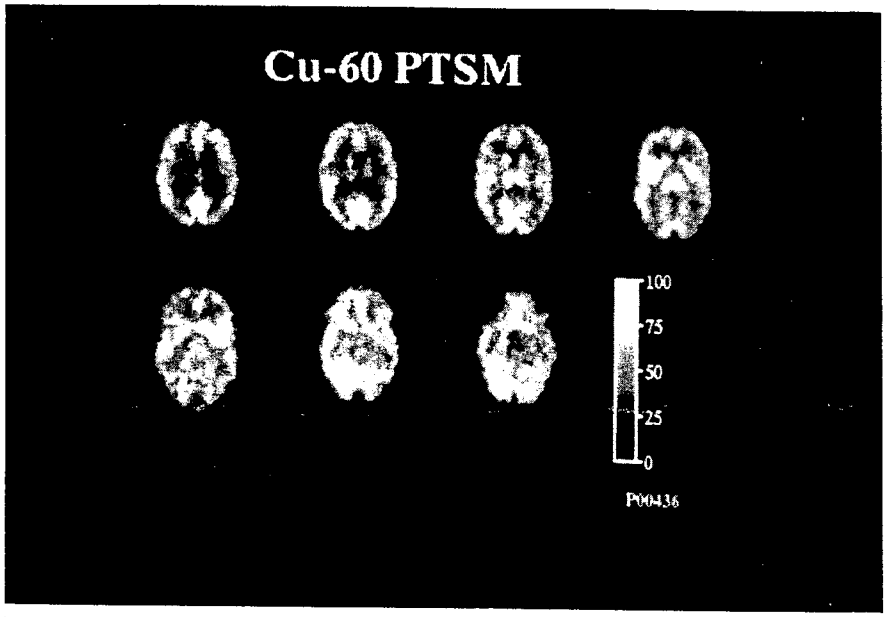


Figure 12.25 **PET image** The trapping of ^{60}Cu-PTSM (a thiosemicarbazone) reflects regional blood flow, modulated by a nonunity extraction into the tissue. (Photo courtesy of Dr. R. J. Nickles, University of Wisconsin.)

image in the time required and small enough to minimize patient exposure to radiation.

The advantages of PET over conventional nuclear imaging include the clarity of the cross-sectional views and the availability of positron emitters that can be compounded as metabolites. It is possible to map metabolic activity in the brain by using tagged compounds to observe uptake and clearance.

The measured quantity in PET imaging is the concentration, in tissue, of the positron emitter. To obtain the actual concentration, it is necessary to calibrate and measure the performance of the machine. Because knowing the actual concentration (in μCi/ml) in the patient may not be so important as knowing the fraction taken up in a particular region, the PET camera can be used to measure the tissue concentration in arbitrary units. A short time after the imaging procedure, a sample of the patient's blood may be placed in a well counter (a scintillation counter) to obtain a reference value. Comparison of the tissue and blood activity yields the ratio of isotope uptake. For example, the local cerebral blood volume and the distribution of activity are measured this way. Because the brain adjusts uptake as a function of the use of various metabolites, brain activity can be measured. A rapid sequence of brain images shows the response of the brain to various stimuli and pinpoints areas of abnormal activity.

As different parts of the brain respond to different stimuli, the PET image shows this activity (Figure 12.25). Normal brains generate one image of brain

activity, but abnormal functioning, tumors, seizure, and other anomalies may also be clearly visible in the map of activity. The PET image of the brain shows the patient's responses to noise, illumination, changes in mental concentration, and other activities. One method of introducing a suitable isotope for brain imaging is for the patient to breathe air containing CO made with ^{11}C. Such short-lived positron-emitting isotopes do not occur in nature but can be created in a small cyclotron. This is done by introducing into the nucleus a proton that, in turn, emits an alpha particle or neutron. For certain elements, the nucleus becomes unstable and emits a positron in a short time. For example, ^{11}C is prepared in a small cyclotron and has a half-life of 20.4 min. The short half-life means that the isotope must be prepared near the point of use. The radionuclide decays rapidly, exhibits high activity during the time necessary to obtain images, and clears the patient in a short time. Clearance is a function of both the radioactive decay and the biological excretion of the material.

12.13 ULTRASONOGRAPHY

We know from old war movies that pulses of sound waves are used to detect submarines. Because wavelength λ, frequency f, and velocity u are related ($u = f\lambda$), it is easy to show that wavelengths in the audible spectrum are only a small fraction of the length of a submarine. A phase change of less than one cycle (360°) would result in a maximum error of position equal to the wavelength: $\lambda = u/f$. To find the error of position of a detected submarine, substitute the velocity of sound in water (1480 m/s) and let f = 1 kHz and λ = 1.48 m. This precision would be adequate for detecting submarines but not for, say, visualizing a human fetus. To obtain precision of 1.48 mm, the frequency of the pulse would have to be increased to 1.0 MHz in the ultrasonic range.

Sound and ultrasound follow rules of propagation and reflection similar to those that govern electric signals. A transmission line must be terminated in its characteristic impedance to avoid reflections. The acoustic impedance Z is a fundamental property of matter and is related to the density ρ and the velocity of sound u: $Z = \rho u$. The fraction of energy R reflected at the normal interface of two different tissue types is

$$R = \left[\frac{(Z_2 - Z_1)}{(Z_2 + Z_1)}\right]^2$$

and the impedances are those of the tissues on either side of the interface (Goldstein, 1988).

Acoustic signals diminish as a function of distance, geometry, and attenuation. In free space, the signals decrease as a result of the inverse square law

Table 12.3 Acoustic Properties of Some Tissues at 1.0 MHz

Tissue	u, m/s	Z, g/(cm²-s)	HVL, cm		R at Interface
Water	1496	1.49×10^5	4100	Air/water	0.999
Fat	1476	1.37×10^5	3.8	Water/fat	0.042
Muscle	1568	1.66×10^5	2.5	Water/muscle	0.054
Brain	1521	1.58×10^5	2.5	Water/brain	0.029
Bone	3360	6.20×10^5	0.23	Water/bone	0.614
Air	331	4.13	1.1	Tissue/air	0.999

because the energy per unit area is a function of the total area of the imaginary sphere at distance r. The signals also decrease as a result of attenuation by the medium. Where α is the coefficient of attenuation and I_0 is the incident signal intensity, the signal intensity is

$$I = \frac{I_0 e^{-\alpha r}}{r^2}$$

When α is large compared to r, the exponential term dominates, and it is convenient to define the thickness of material where the attenuation of the medium decreases the signal by half (the half-value layer or HVL) independently of the geometrical effects. Table 12.3 lists the HVL for water and some tissues. Note that water is not very "lossy" and that the signal decreases 50% for 41 m of water. However, a 50% decrease occurs through only 2.5 cm of muscle. Most biological tissues have high coefficients of attenuation and low HVLs. Attenuation also increases with frequency.

Ultrasound transducers use the piezoelectric properties of ceramics such as barium titanate or similar materials. When stressed, these materials produce a voltage across their electrodes. Similarly, when a voltage pulse is applied, the ceramic deforms. If the applied pulse is short, the ceramic element "rings" at its mechanical resonant frequency. With appropriate electronic circuits, the

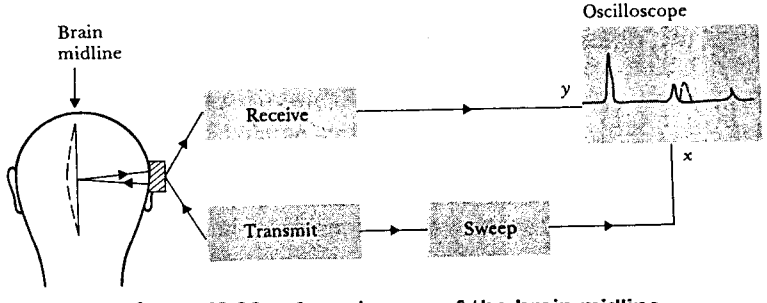


Figure 12.26 A-mode scan of the brain midline

ceramic can be pulsed to transmit a short burst of ultrasonic energy as a miniature loudspeaker and then switched to act as a microphone to receive signals reflected from the interfaces of various tissue types. The gain of the receiver can be varied as a function of time between pulses to compensate for the high attenuation of the tissues. Ultrasonic energy at the levels used for medical imaging appears to cause no harm to tissue, unlike the ionizing radiation of x rays.

The time delay between the transmitted pulse and its echo is a measure of the depth of the tissue interface. Fine structures of tissues (blood vessels, muscle sheaths, and connective tissue) produce extra echoes within "uniform" tissue structures. At each change of tissue type, a reflection results. Figure 12.26 on p. 566 shows how the interfaces of bodily structures produce the echoes that reveal their locations. This type of simple ultrasonic scanner, the A-mode device, was an early device used to measure the displacement of the brain midline. An A-mode device shows echo intensity as an x–y plot. The transducer is placed against the skull and the display gives the echo time of the brain midline (proportional to depth). The transducer is then moved to the other side of the skull and the procedure repeated. The images of normal patients are symmetric so that the brain midline should appear in the same

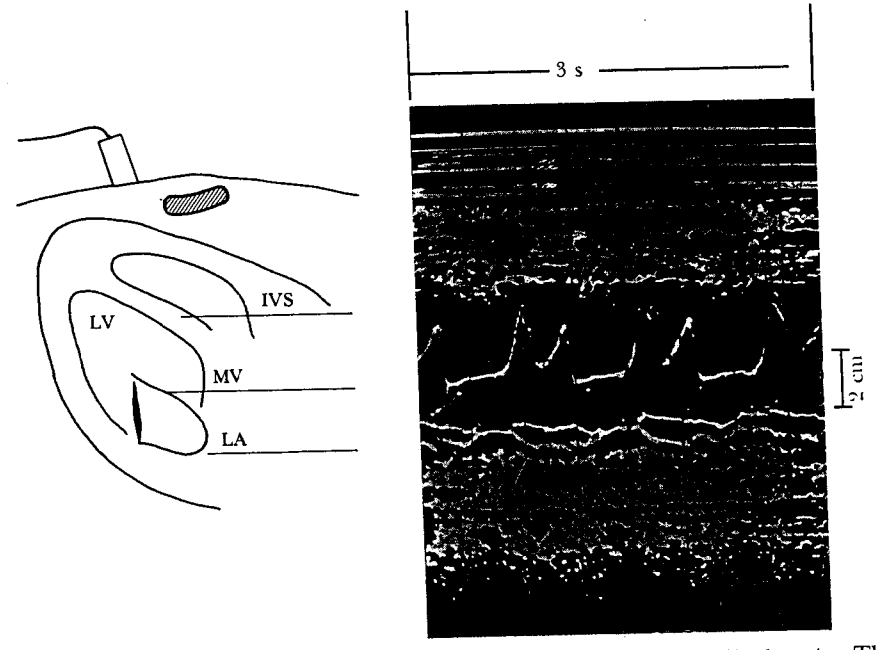
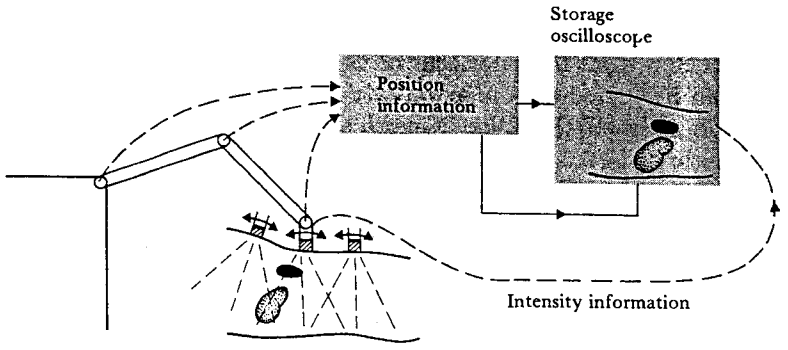


Figure 12.27 Time-motion ultrasound scan of the mitral valve of the heart The central trace follows the motions of the mitral valve (MV) over a 3-s period, encompassing three cardiac cycles. The other traces correspond to other relatively static structures, such as the interventricular septum (IVS) and the walls of the left atrium (LA).



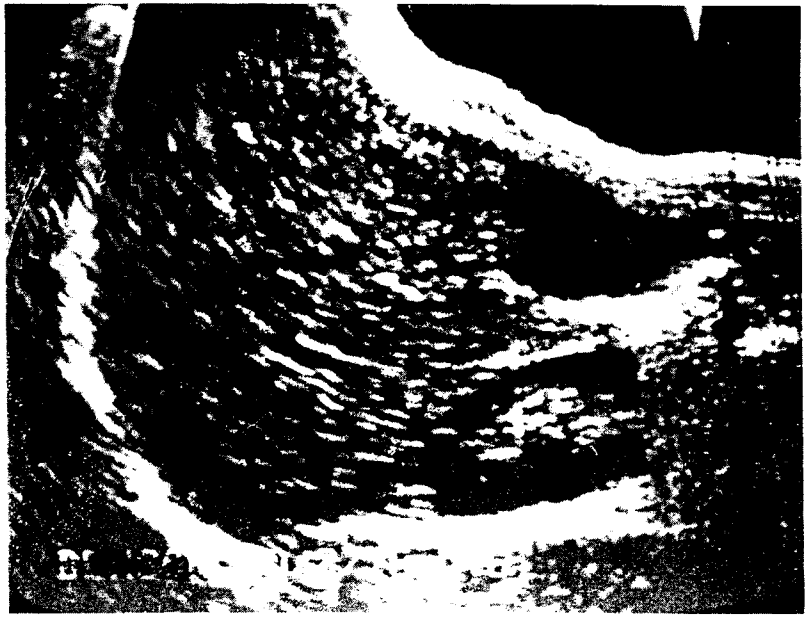


Figure 12.28 (a) B-mode ultrasonic imaging shows the two-dimensional shape and reflectivity of objects by using multiple-scan paths. (b) This B-mode ultrasonic image, which corresponds to (a), shows the skin of the belly at the top right, the liver at the left center, the gall bladder at the right above center, and the kidney at the right below center. The bright areas within the kidney are the collecting ducts.

position in the two images. A tumor or large blood clot could move the cerebral hemispheres to shift the midline. This type of simple device is now seldom used and has been replaced by more elegant systems which show far more detailed structures as well as the brain midline.

If the strength of the echo signal is used to modulate the intensity of the display against the echo-return time, with zero time at the top of the display and with the appropriate image storage circuits, the image can be shown as moving to the right. This technique presents the position of tissues as a function of time as a time-motion or TM scan. Figure 12.27 on p. 567 shows the motion of the mitral valve of the heart over three cardiac cycles.

Computer image storage displays or long persistence phosphor display screens can be intensity modulated as the position of the transducer is varied. The display will show the two-dimensional shape of objects. For older systems, the position and direction of the transducer are coupled to the display circuits by a system of pulleys and potentiometers. Newer systems use mechanical scanning or phased arrays within the transducer assembly and display the two-dimensional image relative to the fixed position of the transducer assembly. A computer stores the echo signals for display as a sector. Some sophisticated systems correlate sectors taken from a number of directions and display them as a single image, much improved in quality over an image taken from only one direction. Sector images and most two-dimensional images are called B-mode images (see Figure 12.28 on p. 568).

Frequencies from 1.0 MHz to 15 MHz are used for most medical ultrasonography. The operating frequency is chosen to meet the imaging task. Higher frequencies will improve resolution, but increased HVL limits the depth of

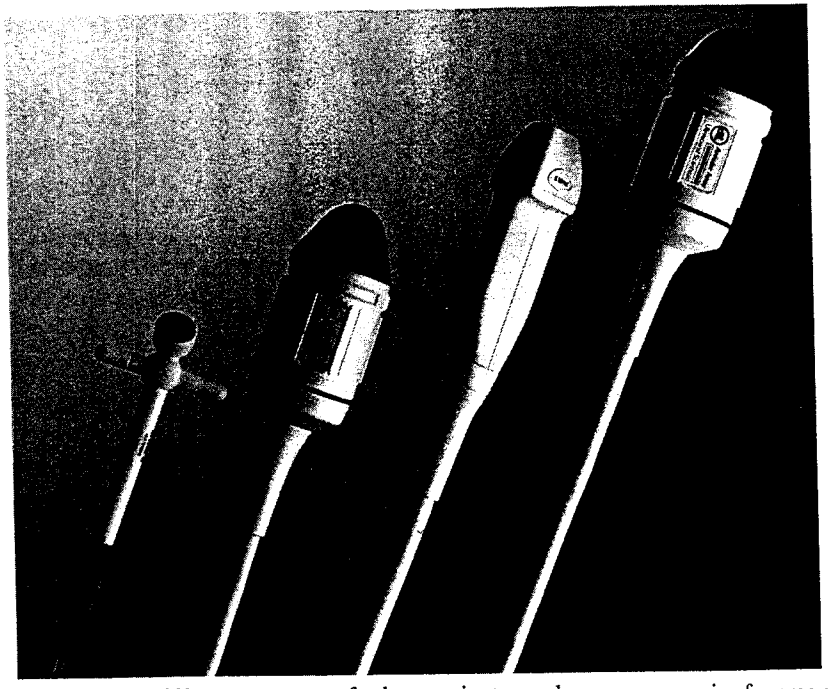


Figure 12.29 Different types of ultrasonic transducers range in frequency from 12 MHz for ophthalmic devices to 4 MHz for transducers equipped with a spinning head. (Photo courtesy of ATL.)

penetration. However, for special purposes (e.g., ophthalmic and neonatal imaging) where the objects are small, the operating frequency may be increased to 15 MHz or higher and the resolution will permit the observation of very small structures or anomalies.

Figure 12.29 on page 569 shows four ultrasonic transducers. The two larger devices use three transducers spinning in fluid-filled enclosures [Figure 12.30(a)]. The midsized device and the smaller, ophthalmic transducer, use phased arrays which can be steered by adjusting the timing of the pulses applied to sets of several ceramic elements. If all elements of the transducer are pulsed at the same instant, the ultrasonic energy will be projected in the forward direction [Figure 12.30(b)]. If the left side elements are pulsed in a delayed sequence, the energy will be projected toward the right with the angle proportional to the timing delay [Figure 12.30(d)]. By adjustment of the timing

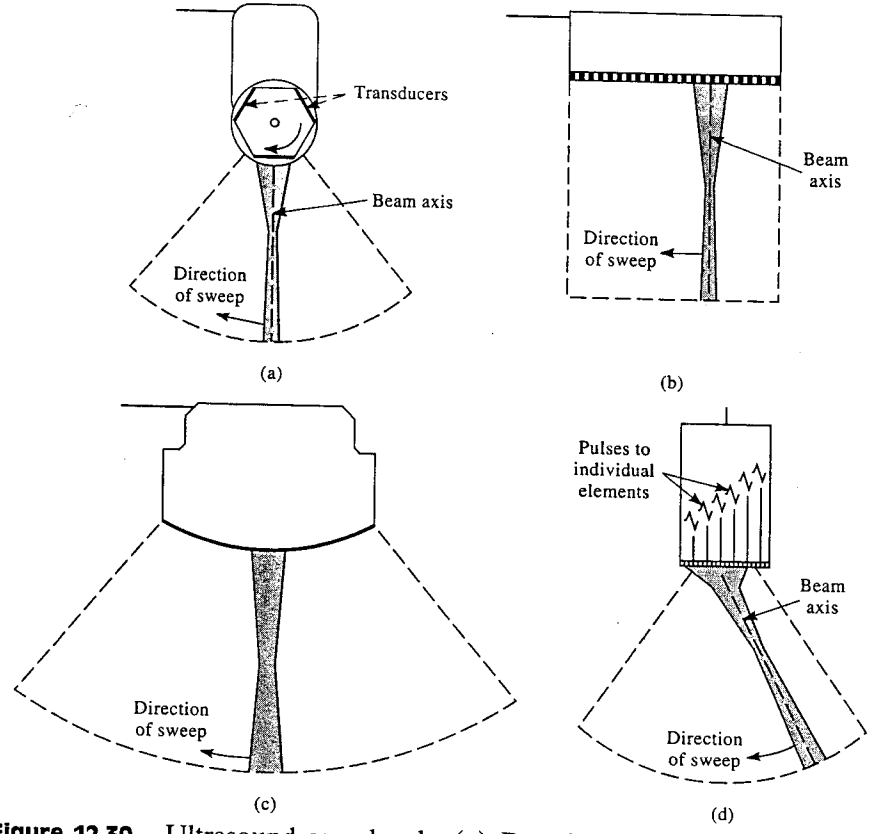


Figure 12.30 Ultrasound scan heads. (a) Rotating mechanical device. (b) Linear phased array which scans an area of the same width as the scan head. (c) Curved linear array can sweep a sector. (d) Phasing the excitation of the crystals can steer the beam so that a small transducer can sweep a large area.

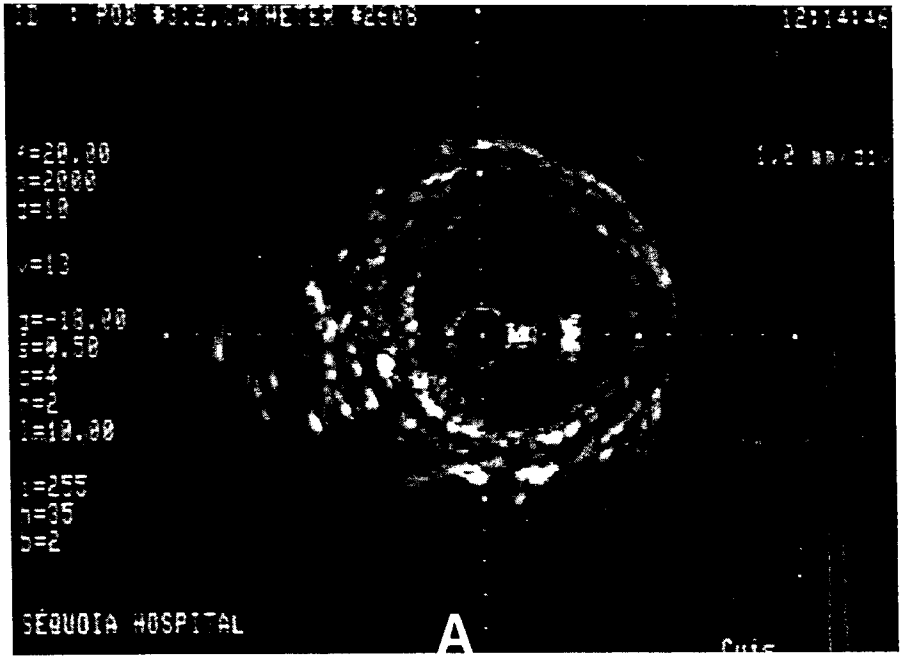


Figure 12.31 Intravascular ultrasonic image showing the characteristic three-layer appearance of a normal artery. Mild plaque and calcification can be observed at 7 o'clock. (Photo courtesy of Cardiovascular Imaging Systems, Inc.)

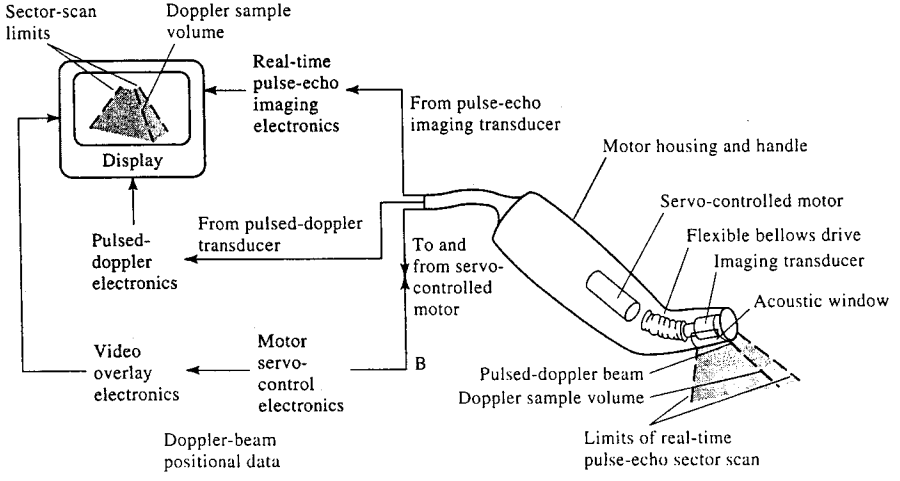


Figure 12.32 The duplex scanner contains a mechanical real-time sector scanner that generates a fan-shaped two-dimensional pulse-echo image. Signals from a selected range along a selected path are processed by pulsed Doppler electronics to yield blood velocity (from Wells, 1984).

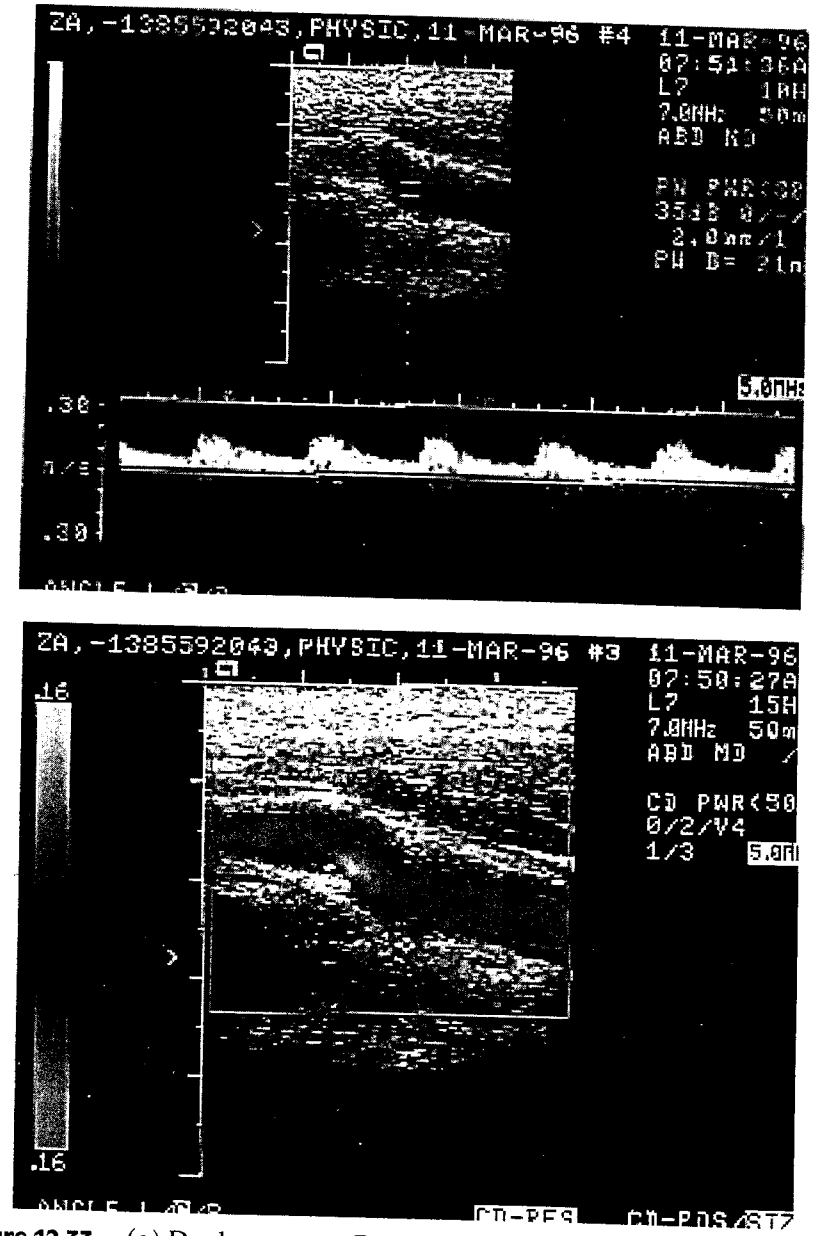


Figure 12.33 (a) Duplex scanner B-mode image and Doppler spectral analysis record for a normal carotid artery, near the bifurcation. The Doppler signals were recorded from the sample volume defined by the Doppler cursor, the two parallel lines located inside the carotid artery. (b) Color flow image of the vessel in (a). Higher velocity components (lighter color, reproduced here in black and white) are seen where the vessel direction courses more directly toward the transducer.

delay, the beam may be scanned from side to side. With either the spinning or phased array transducer placed against the skin, the image of a sector is displayed. Phased array transducers have been made small enough to be mounted at the ends of probes for insertion into body cavities such as the rectum for imaging the prostate or the vagina for showing the fetus or the condition of the reproductive organs.

Even smaller transducers have been made for high-frequency operation. These have been fitted at the tips of catheters and used for examining the characteristics of blood vessels prior to angioplasty. Figure 12.31 on p. 571 shows the appearance of a normal artery taken with a catheter tip transducer. In angioplasty, a balloon is introduced into ischemic or partially closed vessels and then inflated to stretch the walls of the vessel to increase the lumen or diameter and increase blood flow. If the vessel walls are weak, the probe images may show that angioplasty could jeopardize the life of the patient. Following balloon inflation, the probe can be pulled back to determine the dimensions of the stretched walls and verify the integrity of the vessel.

DUPLEX SCANNERS

Halberg and Thiele (1986) describe the design of a phased array ultrasonic duplex scanner that combines real-time two-dimensional imaging with the pulsed Doppler method to measure directional blood velocity noninvasively. Figure 12.30(d) shows how a mechanical real-time sector scanner can generate a fan-shaped beam. Figure 12.32 on p. 571 shows the system block diagram. A colored display from a duplex scanner shows flow into or out of the screen as red or blue against a monochrome background, with the intensity of the color approximating the velocity. This technique is called *color flow imaging* and yields images shown in Figure 12.33 on p. 572.

Because the duplex scanner can distinguish between moving blood and stationary soft plaque, it is useful for diagnosing obstruction in diseased carotid arteries. Pulsed Doppler techniques are useful in locating and determining in the heart the direction and extent of abnormal flow, valvular abnormalities, shunt lesions such as patent ductus arteriosis, and ventricular and septal defects.

PROBLEMS

12.1 How much time does it take to transmit a single television frame over telephone lines that have a bandwidth of 3 kHz?

12.2 A computer monitor uses a noninterlaced scheme with a 4:3 aspect ratio (the height is 3/4 of the width) frame rate of 67 frames/s, 480 visible scan lines, 640 pixels visible/line, 10% retrace time for each vertical and horizontal scan. Find the minimum bandwidth required. Find the bandwidth if the system were redesigned for high resolution and changed to 1024 total scan lines with 920 lines visible and with 1280 pixels visible per line (10%

retrace time). How would the bandwidth be affected if the systems were designed for interlaced scanning and the frame rate was 1/2 the field rate (equal to the original frame rate)?

12.3 A square 8×8 checkerboard is to be imaged using a raster scan system. The board is centered in this square image. When it is rotated 45° with respect to the image axes, its corners are just touching the centers of the image boundaries. Describe the horizontal and vertical characteristics of this raster scan and the total cycles/image required to detect all the checkerboard squares.

12.4 For the Poisson probability density distribution, calculate and plot $p(K;m)$ for $K = 0, 1, 2, 3, 4, 5$ and $m = 3$.

12.5 A 100×100 pixel array has an average of 25 photons per pixel. How many picture elements will randomly exceed this average by more than 17 photons?

12.6 How many photons are required to produce a 200×200 cell picture having 6 gray levels?

12.7 Calculate N_e for $S(f) = 1/(1 + 2f)$, where f is in cycles per millimeter.

12.8 Three elements of an optical system have $N_e = 1, 2$, and 4 cycles/mm. Calculate the system N_e.

12.9 A new imaging system was tested for spatial frequency response, and it was observed that the amplitude response was constant from zero to 10 lp/mm and then fell linearly to zero response at 20 lp/mm. How would this system compare to one having a noise-equivalent bandwidth of 15 lp/mm?

12.10 A film having a gamma of 2.0 is exposed to light and shows a density of $D = 1.0$. What increase of light is required to expose the film to a density of 1.30?

12.11 If we make our measurements in the plane of the patient and must see the smallest pixel of about 0.25 mm (about 2 line pairs/mm), the contrast of the object is about 10%, the QDE of the image detector is 50%, the RL is about 3%, the image rate is 10/s, and the total exposure time is 10 min, what is the approximate total incident exposure? This level of exposure could be seen in an interventional radiographic procedure.

12.12 For all other variables fixed, including probability of detection, plot the dimension d of an object versus contrast C for an x-ray image.

12.13 When an x-ray machine is set for a normal film of the abdomen, around 100 kV, the energy distribution of the beam is such that 4% of the beam can penetrate and exit the patient. If we want to see objects 0.5 mm in diameter that can modulate the beam 50%, what level of incident exposure in R is required? What if the size of the object was 0.2 mm and the modulation (contrast) was 10%?

12.14 Radiation requirements are based on the statistical independence of the x-ray photons producing an image. If the SNR required is, say, 100, then 10,000 photons are required, on average, per pixel. If the image is to be digitized to 10 bits, 1024 levels, why is it necessary to have so many photons per pixel?

12.15 The Hounsfield units are a measure of amplitude resolution in computed tomography. If we need to resolve to 1.0% of amplitude, what does this mean in terms of radiation requirements when compared to conventional radiography (3–5% amplitude resolution)? What is the effect of taking thinner slices in CT in terms of surface (of the patient) incident exposure?

12.16 For the ray θk shown in Figure 12.11, estimate and list the value for each nonzero $W_{ij}^{\theta k}$. For ease of calculating, assume that a complete overlap of beam and pixel corresponds to a $W_{ij}^{\theta k} = 1.0$.

12.17 Our measurement for the ray shown in Figure 12.11 yields $I_0/I^{\theta k} = 2.0$. Calculate our best guess for μ_{ij} using the $W_{ij}^{\theta k}$ values from Problem 12.16.

12.18 Assume that the object in Figure 12.12(a) occupies the center square of a 3×3 square array. Assume that it has a density of 1.0 and that all other squares have a density of zero. Sketch the resulting curves for $p(x)$ and $p(y)$, the projection data for the directions normal to the x and y axes. Sketch the square array shown in Figure 12.12(b), and assign a density for each square in the resulting back projection.

12.19 A patient is placed in the strong magnetic field of an MRI imager. The field is 2.0 T, and the blood velocity (blood is an electric conductor) is 10 cm/s. What is the induced voltage gradient across a blood vessel? Could this be harmful?

12.20 In block-diagram form, show the design of a nuclear-medicine pulse-height analyzer. For each random pulse entering it that has an energy between two limits, it should give only one count. Note that for energies greater than both limits, the output pulse of the detector amplifier passes through both limits twice (rising and falling wave).

12.21 For the gamma camera, describe the x- and y-signal contribution from a photomultiplier that is located in the lower left of the detector array.

12.22 A gamma camera has a line-source response function of $k \exp(-2|x|)$, where k is a constant and x is in centimeters. Calculate the transfer function $S(f)$ of the system.

12.23 Draw a block diagram for an A-scan ultrasonic signal amplifier that corrects for ultrasonic attenuation with distance.

12.24 For Figure 12.26, estimate the sweep speed required for a 10×10 cm display. Estimate the maximal rate of repetition.

12.25 Why are ultrasound images of bone structures distorted? Why don't we observe a similar effect with air cavities?

12.26 In problem 12.11, if the geometry of the imaging system were such that the patient's skin was 70 cm from the x-ray source and the input of the detector, an image intensifier tube, was 100 cm from the source, how would this affect the exposure to the skin of the patient (assuming that the image features were measured at the image intensifier)?

12.27 Three measurements are made of the output of a medical x-ray machine using an ion chamber and electrometer (calibrated in mR) located 100 cm away from the focal spot (source) of the x-ray tube and several thin sheets of aluminum placed at the collimator. The machine was set to 80 kVp, 600 mA, and 0.1 s. The output of the machine was 380 mR (0.0 mm Al), 200 mR (3.0 mm Al), and 163 mR (4.0 mm Al). Find the HVL and output in mR/mAs.