Ultrasonic Diagnostic Instruments

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Summary. The underlying physical principles and current limitations of diagnostic ultrasonic instruments are reviewed. Recently developed ultrasonic imaging devices using pulsed-reflected ultrasound are discussed in detail. These instruments transmit short trains of 1.5- to 10-megahertz sound. Echoes reflected from tissue are converted to electrical signals, which are presented on a display device to outline the contour of tissues and organs within the body. The physical resolution of the system is dependent on several design factors in addition to the transmitted sound frequencies. A resolution volume of approximately 1.5 by 3 by 4 millimeters is achieved optimally with commercially available systems operating at 2.25 megahertz. The various instrument designs are described in the context of clinical usage. Because the sound is diffracted, refracted, and reflected, the imaging considerations are different from those of x-ray imaging. Diagnostic devices based on the Doppler principle are distinguished from pulsed-reflected ultrasonic instruments.

Ultrasound is very attractive as a diagnostic method, from both the patient's and the physician's viewpoint. This nonionizing form of radiation is applied at low power levels, and no harmful effects have been found in humans over the 25 years of very active clinical application. The absence of patient discomfort, apparent safety, ease of performance, and measurement accuracy (relative to other methods for getting similar information) make medical ultrasonography nearly ideal for use in humans. The current generation of clinical imaging devices and recently developed prototype instruments give remarkably complete information about the body's structure and function in comparison with the devices available just a few years ago.

Ultrasound generally is defined as vibrations above 20 kilohertz. For adquate resolution in imaging, the sound frequencies used for most diagnostic instruments range from 1 to 10 megahertz. Such ultrahigh-frequency vibrations are generated by piezoelectric transducers that reversibly convert electrical to vibratory mechanical (sound) energy. When a very short electrical impulse deforms the transducer, a set of sound waves is sent forth into the tissue in contact with the transducer. After this short sound burst, the transducer circuitry is switched to act as a receiver for returning sounds or echoes. The short sound pulse propagates through the body. A very small portion of the sound energy is reflected back to the transducer from each interface between tissues having different acoustic impedence and from structures that scatter the sound within the tissue. The ultrasonograph is calibrated to convert the time elapsed from transmission of the sound pulse to reception of each echo into a measurement of the distance from the transducer to each reflecting surface. Speed of sound in biologic soft tissue is assumed to be constant at an average speed of 1540 meters per second (*1*). For each pulse of sound emitted, the reflectivity of the tissue along the line of sound transmission is obtained.

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Displays for Ultrasonic Images

The echoes are displayed on an oscilloscope screen, with the intensity of each echo represented by the brightness of the spot representing the echo; the position of the echo is displayed in the X-Y plane determined by the position of the transducer and the transit time of the acoustic pulse (Fig. 1). The pulse repetition rate is typically 1 kHz so that all the sound transmitted from a single pulse can travel through the thorax or abdomen and return to the transducer before the next pulse. In Fig. 1, a transducer is represented on the surface of the chest near the heart. The transducer is the origin of a single line representing the path of the sound beam, and the dots corresponding to the location of the echo-producing interfaces are appropriately spaced along this line. Since the trace is refreshed 1000 times per second, moving structures are tracked at this rate. Recording the heart provides excellent time resolution of cardiac motion. The brightness mode (B-mode) display may be recorded over time to produce a time-motion (M-mode) tracing. Diagnostic judgments are based on the pattern of motion written by the echoes from each cardiac structure (*2*).

When the direction of the sound beam is monitored by means of an electromechanical locator, a storage oscilloscope or similar display device is used to record the B-mode tracings with proper spatial orientation. Moving the sound beam through the tissues by moving the transducer is termed scanning, and the image built over time and held on the stored display is termed a static B-mode scan. This method is applied mostly for imaging adynamic organs such as liver and kidneys. Several methods are now available for automatic scanning with the sound beam to create an imaging plane. This process is termed dynamic B-mode scanning because the instruments can supply 30 or more complete frames per second. At this frame rate, moving structures within the body can be assessed.

Axial resolution or the ability to separate two reflectors along the direction of sound travel is determined by the pulse width, which is governed by the wavelength and the bandwidth. For the commonly used 2.25-MHz frequency the wavelength is approximately 0.7 millimeter. Axial resolution approximating 0.7 to 1.4 mm (1 or 2 wavelengths) is expected from such devices (Table 1). The lateral spatial resolution in the two directions perpendicular to the axis of sound propagation is determined by beam shape, which is primarily dependent on wavelength and the transducer diameter. The ultrasonic beam remains essentially uniform over a distance termed the near field and then begins to diverge in the far field. The near field is desirable for imaging because the beam is narrow and the energy is concentrated. For the 2.25-MHz transducer (of 6-mm radius), the collimated near field will be only approximately 50 mm. Most structures of interest are beyond this point. The angle of divergence (θ) of the sound beam in the far field is given approximately by θ = wavelength/transducer diameter. For the

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2.25-MHz transducer, the sound beam is approximately twice the original width when it is 40 centimeters from the transducer face. Materials may be fitted to the transducer face to refract the sound waves and to focus or collimate the beam for optimal volumetric resolution. Systems using an array of small transducers to form a large wave front and to shape the sound beam will be considered later.

The returning echo is converted to an electrical signal that is processed before it is displayed. This processing forms an envelope of the rectified high-frequency signal. In some cases differentiation is used to emphasize the leading edge of this envelope. Many other combinations of standard electronic processing may be used. Receivers must be capable of dealing with signals ranging over 50 decibels (100,000 times in power). Amplifiers of a nonlinear logarithmic type are used to compress this range for display purposes. The strong echoes are from specular reflectors such as the dense and smooth-walled blood vessels at their interface with the blood. Low-intensity signals come from sound scatterers throughout the substance of tissue such as liver or heart muscle.

The earliest displays for ultrasonic images had image storage tubes that were capable of little more than binary or on/off presentations; as a result, clinicians had to be content with mere outlines of the boundaries of various organs and lesions. The next generation of storage devices had electronic storage tubes with dynamic range capabilities comparable to those of the television display. These devices ushered in the era of gray-scale ultrasound in which the large dynamic range of the echoes are compressed to about 30 dB for display; these storage tubes have provided display of diagnostically valuable information about the reflectivity of tissue within the organ. We are now entering an era of digital storage systems whose capabilities in dynamic range and resolution are limited only by the economics of the hardware. The number of picture elements and number of bits per element can be arbitrarily chosen and will presumably grow as these devices continue to become more abundant and less expensive. In addition to the large dynamic range, digital storage systems have the potential for performing complex operations to manipulate the image in ways not available now. These include deconvolution in two dimensions to compensate for the beam spreading and resultant loss in resolution. Also, a variety of more sophisticated schemes to compress the dynamic range of returning signals can be used; the visibility of a narrow range of low-intensity signals can be selectively enhanced, or alternatively, a broad range from low- to high-intensity signals can be displayed.

Table 1. Wavelengths of frequencies used for diagnosis (velocity, 1540 meters per second).

Frequency (MHz)	Wavelength (mm)
1.50	1.03
2.25	0.68
3.50	0.44
5.00	0.31
7.50	0.21
10.00	0.15

Pulsed-Reflected Ultrasonics Instruments

Ultrasonic diagnostic instruments are employed most commonly to make two-dimensional tomographic pictures of the interior of the body. The principle of static B-mode scanning has been mentioned and is widely used but will not be discussed here in detail (Fig. 1). Dynamic scanners can be categorized as mechanical scanners, linear array scanners, and phased array scanners. Mechanical scanners produce their image by rapidly oscillating a single transducer about a fixed point. The transducer is placed on the skin surface, where it is rapidly rocked through a small arc by means of

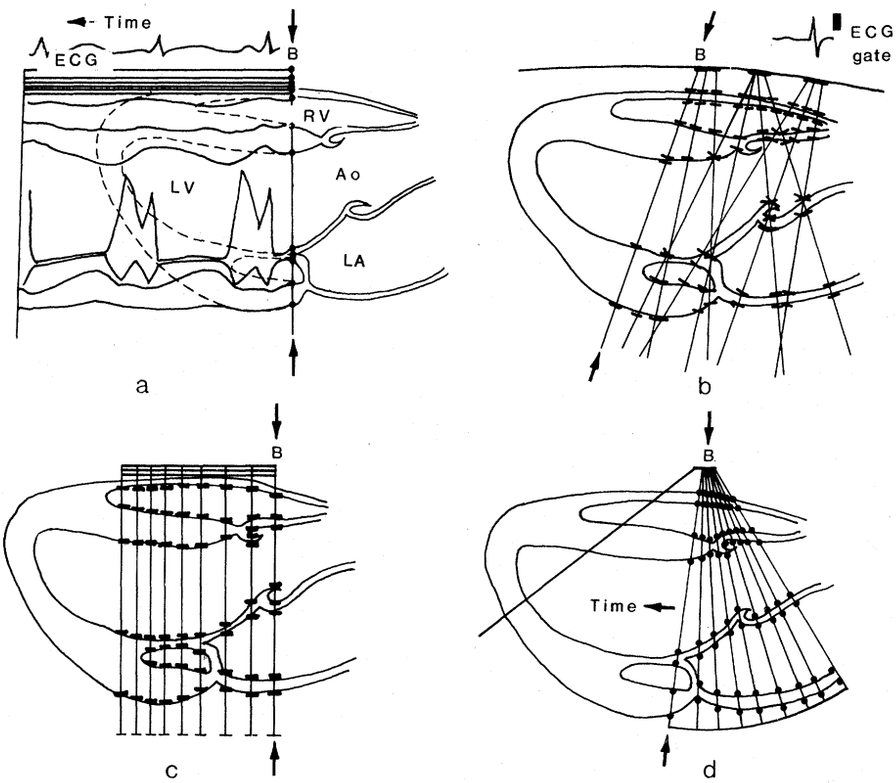

Fig. 1. Commonly used forms of ultrasonic diagnostic instruments. (a) M-mode echocardiography. The sound-transmitting transducer is placed on the chest wall and directed through the heart. The path of the sound beam is represented by a line on a display device showing the brightness or B-mode display (*B*, arrows) of the reflecting surfaces. Each of the reflecting interfaces produces a brightness-modulated signal on the display device, and movement of these structures is recorded over time. The permanent record is then analyzed. (b) For the static B-mode scan, the transducer is moved from place to place over the skin surface, and the sound beam is directed into the tissue to create several paths within the body. The display device holds each B-mode presentation (*B*, arrows) in order to build up an image of the underlying organs. For the heart or other moving organ, the signal may be displayed within a certain time gate. Such an image may be built over many seconds. (c) Linear array, dynamic B-mode scan. The transducer containing a linear array of transmitting crystals is placed on the skin surface (here over the heart), and each individual crystal, or set of crystals, is activated in sequence, with the B-mode presentation aligned on the display device relative to the sequence of crystal activation. Such images can be made at 30 to 160 frames per second with preservation of the dynamics of the underlying tissue. (d) Dynamic B-mode scanning with a sector scan format. The transducer is placed at a fixed point on the skin surface, and the sound path is scanned through a sector of a circular plane. Sequential positions of the sound beam as shown on the display device preserve the spatial orientation of the sound beam path through the tissue. This operation can be performed by use of a mechanical system or by electronic phasing of the transducer array to scan the sound beam path (as explained in the text). Such systems produce 15 to 60 frames per second and preserve the dynamics of motion in the underlying tissue. Abbreviations: *Ao*, aorta; *LA*, left atrium; *LV*, left ventricle; *RV*, right ventricle; *ECG*, electrocardiogram.

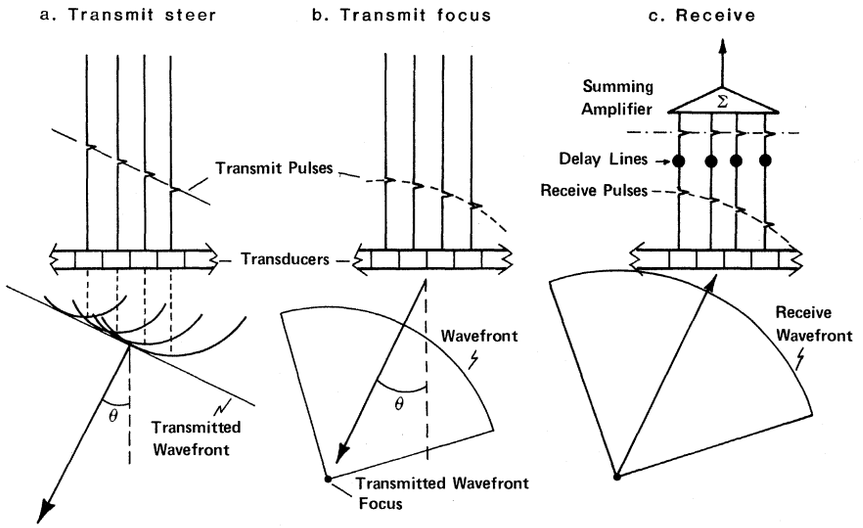

Fig. 2. Principles used in phased array electronic scanning. (a) A sequence of electronic pulses moves toward the transducers and the resulting wave fronts sum to produce a wave front of maximum sound intensity, which is steered into the tissue at angle θ with respect to the direction of the primary sound path. (b) Method used to focus the transmitted wave front in an electronic phased array system. (c) The system used to selectively focus at a given point within the tissue during reception of echoes (*7*) (see text for explanation).

an electrical motor contained within the transducer assembly. The individual B-mode lines are arranged as radii, with the transducer position as their common origin. This sector of a circular plane has been called a sector scan. Such systems were originally described by Griffin and Henry (*3*) and by Eggleton (*4*). Commercial instruments using this method produce approximately 30 images per second. The original systems scanned through a relatively small angle (30° to 45°) because of the mechanical problems of this oscillating transducer maintaining skin contact. Modern commercial systems still use this approach, but many have encased a series of rotating transducers within a housing so that the active transducer could sweep through angles of 90° or more without loss of skin contact.

Linear array scanners contain a number of small transducers arranged side by side, with each element or small group of elements acting as an individual transmitter and echo receiver. These are considered one-dimensional arrays. The B-mode information from each transducer is displayed on a line corresponding to the transducer's position in the array. This produces a rectangular image format as wide as the transducer (*5*). The desire for complete skin contact limits the size of such arrays, although water-containing vessels can be placed between the transducer and the skin to aid this contact problem. The electronic switching from one transducer to another permits very high frame rates, since the frame time is the number of elements multiplied by the round-trip propagation time through the tissue. One of the originators of such systems, N. Bom, introduced the concept of obtaining an M-mode recording from any single B-mode line within the two-dimensional display. This ability to recognize the location of the M-mode line in the two-dimensional projection of the anatomy has been extremely helpful for improving the understanding of the M-mode records. Quantitation is simplified by the use of M-mode data, and these data preserve a record of dynamics on paper. The small size of the individual transducer elements in a linear array system results in rather poor lateral resolution because of diffraction spreading. Current linear array systems use groups of elements to solve this problem; some devices combine the technique of electronically pulsing the individual elements with that of shaping the resulting sound beam by selectively matching the times of activation of neighboring transducers (*6*). This is the principle used in phased array sector scanners.

Phased array scanners use all of the array elements to direct the resulting wave front and thus direct the beam through a sector of a circular plane (Fig. 2). Here all elements are used to produce each of the B-mode lines that constitute the final two-dimensional image. The method of creating and directing the sound beam is as follows. A single transducer is activated to produce its wave front that begins to advance into the tissue. A very short but carefully calculated time later, the next element is activated to produce its acoustic pulse. As this process continues, the individual acoustic wave fronts combine to produce a maximum acoustic intensity along a direction determined by the intersection of each wave front. The direction of this maximum acoustic intensity, or newly constructed wave front, is governed by the time of activation of each element. This procedure is equivalent to tilting the transducer array to direct the beam. The acoustic beam may be directed along any path into the tissue by changing the sequence of activation. The beam may be further altered by including some focusing of the transmitted beam; this is done by combining a curved timing sequence with a linear one to produce a resultant beam that is focused at a given distance and directed in an arbitrary angle (Fig. 2). This principle, which was extensively explored by von Ramm and Thurstone (*7*), can be used in the reception of echoes in a more general sense; the receiver focus can be synchronized to the range of returning echoes in a dynamic focusing system to improve the resolution of structures at all feasible depths.

The phased array systems used thus far are capable of dynamic focusing only within the plane of the cross section. To minimize the thickness of the section it would be optimum to focus in the direction normal to the scanning plane. Theoretically, it is possible to achieve focusing in both lateral directions by the use of a series of linear arrays placed side by side (two-dimensional arrays), although at present this has proved somewhat impractical. Modified versions of two-dimensional arrays are used by some commercial manufacturers to improve the resolution normal to the plane. Most commercial instruments with phased array systems include fixed-focus lenses that provide improvement in beam pattern over a limited depth range (*8*). A concentric annular ring array of transducers, with a controlled delay element for each ring, is a good approach to dynamic focusing in both lateral dimensions. Unfortunately, this configuration is incapable of deflecting the beam to form a sector scan. If the anatomy allows the transducer to be displaced from the skin by a water-containing vessel, then an annular array system may be used with mechanical devices, like mirrors, to deflect the beam. A unique approach to focusing in the two dimensions perpendicular to sound propagation is to focus in one dimension (such as the vertical dimension) on transmission and then to focus in the orthogonal (horizontal) dimension during reception. The difficulty with this approach is that electronic dynamic focus-

ing is available only in the receiving mode, since once a wave front is launched, all control over it is lost. The annular ring array is the only configuration devised thus far that maintains its focus during propagation. It is possible to use an appropriately weighted annular ring array for transmission to provide diffraction-limited vertical resolution, and to use a dynamically focused linear phased array in the conventional fashion to provide optimal horizontal resolution for reception of the signal (*9*). The combination is known as the theta array because the Greek letter resembles the linear array within the annular ring. Initial experiments with this system appear promising. The use of relatively large two-dimensional arrays is particularly desirable in abdominal imaging, where the anatomy does not significantly limit the size of the aperture. Arrays of 5 by 5 to 10 by 10 cm can provide resolutions of less than 1.0 mm in all dimensions.

Any of the electronic systems can give simultaneous cross-sectional dynamic images and in addition provide one or more M-mode records of individual lines. Electronic switching of these systems permits a higher sampling frequency from the individual M-mode line than is provided by the frame rate of the two-dimensional image. Although there are other systems using variations of these basic forms, most commercial instruments now in use are either dynamic B-mode scanners of the type described or static B-mode scanners.

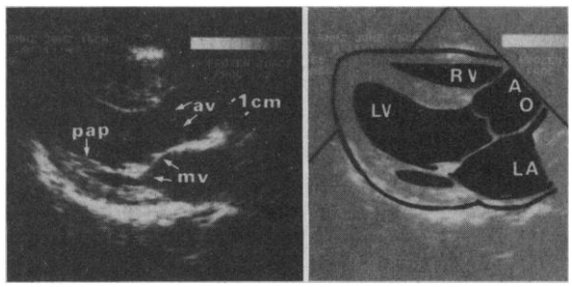

Fig. 3. (Left) A single frame from a phased array dynamic B-mode sector scanner showing a section, parallel with the long axis, of the heart. (Right) Same frame with diagrammatic overlay added. Abbreviations: *av*, aortic valve; *mv*, mitral valve; *pap*, papillary muscle.

Clinical Applications

When these methods are applied to the heart (echocardiography), frequencies of 2.25 to 5 MHz are used for imaging. As a practical matter, 2.25 MHz gives an axial resolution of approximately 1.4 mm, whereas the lateral resolution of typical phased array systems is approximately 3 mm at the nominal focal distance. The resolution perpendicular to the plane of scan at this distance is usually somewhat greater—typically 5 to 6 mm—because beam divergence increases with distance. Nevertheless, these numbers give some idea of the accuracy of measurement of the internal dimensions of cardiac chambers. Studies in which measurements of cardiac wall thickness are compared with direct measurement at autopsy have shown an accuracy of 1 to 2 mm. Serial echocardiographic studies have shown excellent reproducibility of the measurements (*10*). The two-dimensional images clearly represent the anatomy and dynamic changes in the heart to those who have learned basic anatomy and physiology (Fig. 3). Cyclic variations in chamber size and wall thickness conform to the well-known physiology of these structures.

Localized abnormalities of wall motion and wall thickening within the left heart are well identified on ultrasonic images. Details of internal cardiac anatomy are displayed so that the operator may choose any arbitrary plane within the heart to make measurements. We use our knowledge of anatomy and physiology to choose an appropriate plane for specific purposes. For example, we may wish to measure the minimum area of orifice of the heart valve to determine the adequacy of this opening for blood flow. Measurements of the area of the mitral valve by two-dimensional echocardiography have shown good correlations with calculations of the dimensions of the mitral valve orifice made in the conventional way with hemodynamic measurements (*10, 11*). During heart catheterization, hollow tubes are passed through the blood vessels to the heart to measure the pressure drop across a valve and to measure flow. The dimensions of the valve orifice are routinely calculated from these parameters. Measurements made by two-dimensional echocardiography match the measurements made at catheterization very well and have the advantage of displaying the actual orifice rather than deriving its dimensions from measurements of pressure and flow. Many physicians choose to use ultrasonic imaging as a substitute for heart catheterization, both for diagnosis and for determining the severity of the valve disease. In measuring such small structures, a great deal of operator skill is needed to obtain consistent results. The detection of very small masses of abnormal tissue within the heart, such as clots, masses of infecting organisms, or intracardiac tumors, is better done with echocardiography than with any other diagnostic method.

Children with congenital heart disease may have very complex interrelations of the cardiac chambers and vessels; some of the structures may be missing altogether or there may be abnormal communications between them. The unique ability of ultrasound to provide detailed anatomy of the interior of the heart permits the experienced echocardiographer to visualize the anatomy. A series of parallel or orthogonal tomograms can be assembled into a three-dimensional image of the organ. Ultrasonic imaging truly has revolutionized the recognition and understanding of congenital cardiac anomalies.

To give a few other examples, current instruments have sufficient resolution to detect the human ovum 4 to 6 weeks after conception. Details of fetal growth can be monitored with repeated studies because of the accuracy and safety of this technique, and a large number of congenital abnormalities can be detected early in gestation. Normal growth and development patterns and accurate fetal age are best predicted by measurement of fetal head size and fetal volume, both of which are most accurately available through these ultrasonic methods. The details of internal architecture of various abdominal organs are so well presented with modern instruments that one can detect very subtle abnormalities of liver, pancreas, or kidneys (Fig. 4). Ultrasonic imaging is used widely to detect tumor nodules within the liver, enlarged or abnormal lymph nodes in the abdomen, small stones within the gallbladder, and localized pockets of infection or fluid accumulation.

Comparison with Other Methods

It is instructive to compare ultrasound and x-ray systems in their respective roles in medical imaging. As indicated, ultrasound is free from toxicity at the levels used, whereas x-rays are an ionizing form of radiation and have the potential to be damaging. In fundamental physical differences, x-rays propagate at the speed of light, 3×10^8 meters per second, whereas ultrasound waves travel in tissue at approximately 1500 meters per second. This large difference makes direct acquisition of three-dimensional

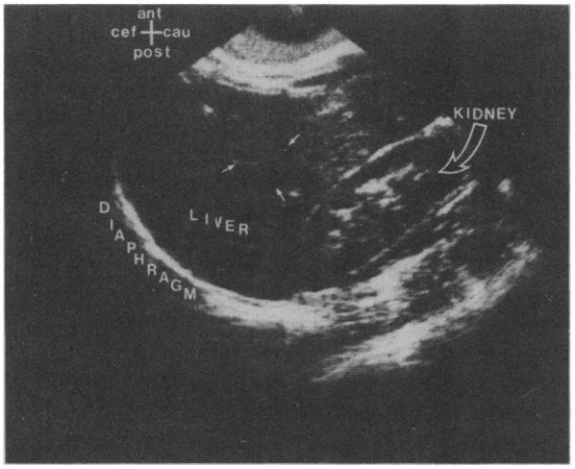

Fig. 4. Abdominal sonogram showing the liver, kidney, and diaphragm in sagittal section. Small arrows indicate blood vessels within the liver. Abbreviations: *ant*, anterior; *cau*, caudad; *cef*, cephalad; *post*, posterior.

data possible with pulse echo systems; x-ray systems are not capable of providing direct acquisition of three-dimensional data. In x-ray systems, the sum of the incremental attenuation coefficients is obtained along the ray path. Although computed tomography systems can be used to reconstruct the local attenuation coefficient, this requires measurements at all angles. Thus the individual local values of attenuation cannot be directly acquired with a single measurement. In ultrasound, however, the local value of reflectivity is given directly by a single measurement.

Ultrasound requires a soft-tissue path between the transducer and the region of interest, and intervening bone or air results in severe attenuation and distortion of the sound, whereas x-rays can propagate through all materials of interest. In addition, x-rays travel in straight lines and do not exhibit diffraction, refraction, or reflection, potentially distorting effects that are present in ultrasound. Since the velocity of propagation of sound is very nearly the same in most soft tissues, refractive effects are minimal for ultrasound. Diffraction, because it results in the divergence of ultrasonic beams, represents the most serious limitation to resolution; these effects are best counteracted by focusing (discussed above). The properties of diffraction, refraction, and reflection allow ultrasonic systems a great deal of flexibility since, unlike x-rays, the energy can be deflected and focused into a region of interest (*12*). Thus, properties that can cause distortion of the propagating waves allow them to be manipulated into desired configurations. Ultrasonic imaging systems operating in the pulsed-echo mode do not directly measure some available physical parameters of tissue; the image produced corresponds to the local reflectivity of the tissue, based on local changes in acoustic impedance and often on the relative angle between the tissue interface and the ultrasonic beam. However, ultrasound can be used in the transmission mode to provide cross-sectional images of the acoustic velocity and attenuation of sound. In a direct analogy to x-ray computed tomography, images of the exised heart and of the breast in vivo have been made by Greenleaf *et al.* (*13, 14*). As with x-ray computed tomography, transmission measurements are made for an array of positions with a complete set of angular increments. For each path, the time of flight (the time of transmission through the tissues) and the amplitude of the received signal are measured with a transmitting transducer on one side of the subject and a receiving transducer on the other side. The time of flight represents the sum of the incremental velocities along the ray path, and the amplitude represents the sum of the incremental attenuation values. As in the methods used with classical computed tomography, these values are used to reconstruct cross-sectional images of the velocity and attenuation. These physical measurements are often well correlated with disease processes, as shown in breast tissue (*13, 14*). Ultrasonic computed tomography systems present a number of practical difficulties. Since transmission measurements are made at all angles, the method is essentially limited to breast studies. Also, refractive effects cause the beam to travel in other than straight paths; this error is difficult to compensate for. Mueller *et al.* (*15*) have been studying a similar approach in which a transmission measurement is made over the entire extent of the object at each angle. The resulting projection at each angle includes the effects of refraction and is therefore more accurate, although considerable computation is required.

Doppler Techniques

Specially designed ultrasonic systems have the capability of measuring the flow velocity of moving structures, such as blood cells within blood vessels, by comparing the transmitted sound frequency with the Doppler shifted frequency of the returning sound. The frequency shift is proportional to the velocity of flow. For velocity measurements, the systems use continuous transmission of sound and a second crystal receiver; the resultant frequency shifts are due to all flow taking place throughout the beam path. Pulsed range-gated Doppler systems have been designed to isolate flow in a particular vessel or a portion of a vessel (*16*). In such systems a burst of sound is transmitted as described for echo imaging. The sound returning to the transmitting crystal a finite time after transmission represents echoes returning from a certain depth of interest. The sound burst is usually somewhat longer than that used in imaging since there must be sufficient cycles to estimate frequency accurately. As a general rule, the highest Doppler shift frequency that can be measured is half of the pulse repetition rate. Thus deep vessels that require long round-trip transit times need reduced pulse repetition rates, and there is greater ambiguity in measurements of flow from such vessels. Circuitry to differentiate positive from negative Doppler shifts allows differentiation of forward or backward flow relative to the sound path. In applying these systems clinically, it is necessary only to compress the patient's calf muscle or thigh muscle while observing flow in the venous system to tell whether there is flow within veins and whether this flow is antigrade or retrograde. Range-gated directional Doppler systems can be used to observe the physiology of flow in virtually any vascular system the sound can penetrate. There are various approaches for estimating average flow within a vessel and for combining this with information about the size of the vessel, so that volume flow can be calculated. Two-dimensional imaging systems and Doppler systems of this type have been joined to allow the diagnostician improved accuracy in positioning the volume for sampling Doppler shift in frequency (*17*).

Areas for Future Improvement

Although the ultrasonic images now obtained for diagnoses have far surpassed those available or even anticipated a few years ago, we would still like

to have perfect images in each patient. There are several physical obstacles that are difficult to overcome, such as the very poor transmission of sound through air that intervenes between the sound transducer on the body surface and areas of interest within the body. However, other problems appear to be more tractable. These include the appearance of "speckle," problems with inhomogeneous media, and problems with multiple reverberations. Speckle is a grainy pattern, most visible in uniform interior regions of organs such as the liver. Because of the coherent nature of ultrasound, reflections from adjacent scatterers within a small region can interfere constructively or destructively to produce the resulting pattern. This random addition and subtraction of scattered wave fronts creates a mottled appearance that is often interpreted as a property of the tissue involved. Unfortunately, this pattern may obscure a true lesion that is immersed within this noisy environment. There are relatively few studies of this phenomenon although it is now attracting interest. Since the speckle patterns are random, one approach to reducing this effect is to sum a number of images taken from different positions or to use different frequencies to image the same tissue. The resultant speckle will be reduced by the square root of the number of independent images in such a system.

For echo imaging systems, the propagation velocity and attenuation of sound are assumed to be constant. However, variations in velocity and attenuation can seriously distort the sound beam and degrade the image. One possible approach to this problem is the use of variable time-delay devices within phased array systems; the delays could be manipulated to partially compensate for velocity variation and hence to minimize the resultant distortion. Also, images taken from different angles could be combined to minimize distortion.

Echo imaging systems are based on the assumption that the first echo originating from an organ or scatterer is the only echo received with significant intensity. It is assumed that other echoes in this area result from multiple reverberations or reflections of the wave front bouncing back and forth between nearby interfaces. Because the primary reflection is a very small percentage of the transmitted energy, multiple reflections become vanishingly small. However, in a few select cases involving bone or regions containing gas, the reflections are high enough to become visible and must be considered. Clinical operators have learned to recognize and deal with such reverberations by creating images from different transducer positions. Nevertheless, these considerably distort the primary image and may obscure an area of interest.

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Three-Dimensional Imaging of Heart, Lungs, and Circulation

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The viability of humans and animals is critically dependent on virtually continuous motions of blood and gases. These motions are generated by movements of all the individual structural components of the circulatory and respiratory systems. Accurate understanding of the interrelationships of these coordinated movements requires synchronous measurements of the changes in shape, dimensions, and position of an entire anatomic structural system simultaneously with its physiologic functions. An example of such an interrelationship is the change in shape and dimensions of the thorax and lungs and the simultaneous spatial distribution of pulmonary ventilation.

Inability to simultaneously measure these parameters and their interrelationships has limited the insights that can be obtained regarding their integrated physiologic functions in the intact organism. This inability has also limited understanding of the multifaceted mechanisms by which various abnormalities, such as coronary artery disease, affect these functions.

The dynamic spatial reconstructor (DSR) was designed to increase insights into physiologic and pathophysiologic interrelationships between anatomic structural dynamics and the corresponding physiologic functions. It is an x-ray imaging device that operates on the principle of computed tomography (CT) and

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