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Basic Physics and Imaging Characteristics of Ultrasound

George Kossoff, D.Sc. Eng.

1 Bowes Avenue, Killara, New South Wales 2071, Australia

Abstract. The imaging characteristics of diagnostic ultrasound (US) are determined by the ultrasonic properties of tissue. The velocity of propagation of US and the attenuation are the 2 most important parameters. These determine the frequency with which the tissues may be imaged, which in turn sets a fundamental limit on the axial and the lateral resolution. Ultrasonic imaging equipment is designed on the premise that the ultrasonic energy propagates through tissue in a straight line and that the ultrasonic beam is very narrow. In fact, the ultrasonic energy propagates through tissue as a beam of finite dimensions set by the physical dimensions of the transducer, the way it is constructed, and the way it is energized. Also, the velocity of propagation in different tissues varies and this can lead to deviation of the ultrasonic beam from the assumed direction of propagation. This breakdown in assumptions leads to the creation of artifacts that must be appreciated in the interpretation of ultrasonic images. For this reason skilled interpreters of ultrasonic images follow 3 golden rules: never make an interpretation on a single image; just because a feature is displayed do not consider that it is necessarily real; and just because a feature is not displayed do not consider that it is necessarily not there.

Basic Physics

Modes of Propagation

Ultrasound (US) is mechanical energy that propagates through tissue as an oscillating wave of alternating pressure [1, 2]. The elements of the tissue move in response to the applied pressure and this forms the mechanism by which US interacts with tissue. The unit for pressure is a megapascal (MPa). Diagnostic equipment generates pressures ranging between 0.5 and 5 MPa. These are high values compared with the atmospheric pressure of 0.1 MPa.

The direction of vibration of the elements of tissue can be either along or at right angle to the direction of propagation of the ultrasonic wave. The former are known as longitudinal waves, the latter as shear waves. Liquids and soft tissue have resistance to compression but not to shear deformation as witnessed by the fact that liquids flow. For this reason liquids and soft tissue support the propagation of longitudinal waves only. Hard tissue such as bone can support the propagation of longitudinal as well as shear waves.

Velocity of Propagation

The ultrasonic wave propagates through tissue with a velocity that is specific to the tissue. For example, the velocity of propagation of the longitudinal wave in normal saline at body temperature is 1540 m/s. The velocity of propagation in most soft tissue is between 1% and 2% faster, while in fat it is about 10% slower. The velocity of propagation is a parameter whose value does not depend on the frequency.

The velocity of propagation of the longitudinal wave varies between 3000 and 5000 m/s, depending on the composition of the bone. The velocity of the shear wave is typically between 20% and 30% slower than that of the longitudinal wave.

The velocity of propagation in air by comparison is quite slow, 440 m/s. This is one of the several reasons why air-containing structures are not amenable to examination by US.

Wavelength

Wavelength is defined as the distance between 2 corresponding points on a wave. Wavelength is calculated by dividing the velocity of propagation by the frequency. For example, in normal saline the wavelength at 3 MHz is 0.5 mm, whereas at 5 MHz it is 0.3 mm.

Wavelength is of major importance in diagnostic US because it determines the imaging resolution of the equipment. Typically the axial resolution of the equipment ranges between 2 and 4 wavelengths, whereas the lateral resolution ranges between 3 and 10 wavelengths. Thus at 5 MHz, the axial resolution of the equipment ranges between 0.6 and 1.2 mm, whereas the lateral resolution ranges between 0.9 and 3 mm, depending on the quality and the cost of the equipment.

Acoustic Impedance

Acoustic impedance of tissue is a parameter that is defined as the product of the density of the tissue times the velocity of propagation in the tissue. As the density of liquids and of soft tissue does not differ significantly, the acoustic impedance of liquids and soft tissue also does not differ by more than several percent.

When ultrasonic energy propagates from one medium into another, some energy is reflected back while the rest propagates into the second medium. The amount of energy that is reflected back is determined by the difference in the acoustic impedance between the 2 media. Thus in the propagation through liquids and soft tissue, most of the energy continues to propagate deeper into the tissues. The internal structure of soft tissue is heterogeneous and for this reason soft tissues return low level textural echoes.

Liquids, on the other hand, are homogeneous and do not give rise to internal echoes.

A different situation occurs in the propagation from soft tissue into bone. Because of impedance mismatch considerations only about 30% of the energy propagates from soft tissue into bone. A similar impedance mismatch occurs when energy arising from a reflection within bone impinges on its return path on the bone-soft tissue interface. For example, a totally reflecting interface within bone returns only about 10% of the incident energy compared with 100% that such an interface reflects when located in soft tissue. Thus the sensitivity for imaging internal bone structure is reduced significantly.

The density of air is very much less than that of soft tissue. Therefore, the acoustic impedance of air is much smaller than that of soft tissue and >99% of energy is reflected back from a soft tissue-air interface. This is another reason why air-containing structures cannot be examined by US.

Attenuation

As the ultrasonic wave propagates through tissue, the energy content in the beam diminishes progressively. Several mechanisms are responsible for this loss, and the term attenuation is used to describe loss of energy from all causes. Attenuation is frequently subdivided into absorption, which involves transfer of energy from mechanical form into heat, and scattering, which involves dispersion of energy from its main direction of propagation.

Liquids do not significantly absorb or scatter the ultrasonic energy and are generally considered to be nonattenuating. Their attenuation is, however, proportional to the square of the frequency and can become significant at frequencies higher than 15 MHz. Soft tissue is a significant absorber but a relatively mild scatterer of the ultrasonic energy. The attenuation by soft tissue is proportional to frequency and is specified in units of dB/cm/MHz. For example, liver attenuates US at a rate of 0.45 dB/cm/MHz. Thus, a one-way propagation through 5 cm of liver at 5 MHz diminishes the energy content by 11.25 dB, whereas in the “there and back” mode used in imaging the energy content is reduced by 22.5 dB.

Bone is a significant absorber as well as scatterer of US. Also, when energy impinges on bone at angles other than normal, some of the energy is transformed into a shear wave, which is absorbed very rapidly by bone. For this reason it is frequently assumed that all of the energy that penetrates bone is attenuated fully at the front surface of bone.

Air-containing soft tissues absorb as well as scatter the ultrasonic energy. The attenuation is high and the energy is dispersed in all directions throughout tissue. As a consequence, air-containing tissues such as lung are not amenable to examination by US.

Decibel Notation

The decibel notation is used frequently in the description of the imaging characteristics of ultrasonic equipment. It is a descriptor that compares the value of a parameter at the output relative to that at the input.

In case of pressure or voltage, the decibel is defined as 20 times the logarithm of the ratio of the pressure or voltage at the output to that at the input. For example, if the voltage at the output of the equipment is higher than that at the input due to gain of the equipment, the decibel value of gain is a positive number. However, if pressure is considered, and the value of pressure is reduced by attenuation, the decibel value for pressure in this case would be a negative number, indicating a loss of energy. This can cause a less experienced user some confusion as he or she develops familiarity with the notation.

Table 1. Voltage gain and equivalent value in decibels.

Voltage gain	Gain in decibels
1	0
1.3	2
2	6
4	12
10	20
100	40
1000	60
10,000	80

There are a number of advantages to the decibel notation which are based on the mathematical properties of logarithms. For example, if the ultrasonic energy propagates through 2 tissues with different attenuation, the total attenuation is obtained by simply adding the 2 decibel attenuation values. Logarithms compress the magnitude of numbers. For example, the logarithm of 10 is 1, of 100 is 2, of 1000 is 3, etc. In the decibel notation, a voltage gain by a factor of 10 is equivalent to a gain of 20 dB, of 100 to 40 dB, of 1000 to 60 dB, etc. Once mastered the decibel notation is a convenient way of expressing, in small numbers, the large range of values of parameters such as gain and attenuation which are commonly encountered in describing the performance of the ultrasonic imaging equipment.

Table 1 lists some of the commonly quoted voltage gain factors and the equivalent value of gain in decibels. Other values may be derived from those listed, using the relation that the product of the 2 voltage gain factors is the sum of the 2 attenuation values. For example, consider a situation in which the voltage gain of the receiver is 4000. This can be restated as a gain of (4×1000). From Table 1 this is equivalent to ($12 + 60$) dB, i.e., a gain of 72 dB.

Transducer

The ultrasonic energy is generated by a transducer that uses a piezoelectric ceramic to convert an applied electric voltage into an ultrasonic wave that propagates into tissue and reciprocally converts the reflected ultrasonic energy from acoustic impedance mismatches in tissue into a received voltage. The ultrasonic energy emanates from the transducer in a beam pattern that is determined by the dimensions of the transducer, the way it is constructed, the way it is energized, and the frequency.

Single Element Disc Transducer

The simplest and still occasionally used transducer consists of a single element circular disc. The frequency is set by the thickness of the disc, a 5 MHz transducer, typically about 0.4 mm thick.

The ultrasonic wave initially propagates in a cylindrical beam, the diameter of the beam being equal to the diameter of the transducer. The wave propagates in this form for a certain distance known as the near-field. The near-field is proportional to the square of the diameter of the transducer and inversely proportional to the frequency. Beyond the near-field, the beam changes from a cylinder into a cone and diverges into the far-field at a constant angle that is proportional to the diameter of the transducer and inversely proportional to the frequency.

Focusing of the beam to improve the lateral resolution may be achieved by placing a spherical lens in front of the transducer. Diffraction considerations dictate that focusing can only be achieved in the near-field of the transducer. Unfortunately, it is also not possible to narrow the beam over the total near-field and a reduction in beam width is obtained only around the focal zone. If the focus is set at half the near-field distance, the beam at focus is narrowed by a factor of about 5, i.e., the beam width at focus is one fifth the diameter of the transducer and the focal zone centered about the focus is about one fifth the near-field distance. A stronger focus can be obtained by placing the focus closer to the transducer. The focal zone is then also proportionately reduced.

Annular Array Transducer

Annular array transducers have been developed to overcome the beam pattern limitations of single disc transducers. As implied by the name, these transducers consist of a disc transducer surrounded by a set of, typically 7, annular transducers. Focusing of the beam is achieved electronically by energizing the outer ring first, and then progressively the next outer annular transducer until finally the central disc transducer is energized. The position of the focus is determined by the time delay between the energizing pulses. This time delay, like a lens, is generally spherical in form and the beam comes to focus at the center of this sphere. The advantage is that the position of the focus is under electronic control and several different enegizing sequences can be used to generate a narrow beam over the required examination distance.

The beam pattern of a single disc and an annular array transducer is shown in Figure 1a. The focal zone of a single disc transducer is relatively shallow and a much improved beam pattern is obtained with an annular array transducer. As can be seen in Figure 1a, 4 different energizing pulse sequences were used to obtain the annular array beam pattern.

Linear, Curved, and Phased Array Transducers

Linear, curved, and phased array transducers are modern developments in transducer technology. The common design feature is the employment of many, typically of the order of 100, small rectangular transducers to generate the ultrasonic beam.

The beam that emanates from a rectangular transducer can be approximated by considering a corresponding beam emanating from a disc transducer of diameter equal to one of the sides of the rectangular element.

Consider first the beam emanating from the long side of the rectangular element. This dimension is usually similar to that employed by the single element disc transducer. The beam in that plane is therefore similar to that described previously; i.e., there is a near-field distance and the beam may be focused in that plane by a cylindrical lens in front of the transducer.

The short side of the rectangular element is usually 1–2 wavelengths long for linear and curved array transducers. The near-field distance is very close and the beam diverges broadly as it emanates from the short side of the element. Focusing of the beam is achieved electronically by energizing the elements at slightly different times so that the beams from individual elements constructively interfere to generate the desired focused beam. As in the case of the annular array transducer, focusing is achieved by energizing the outer elements of the transducer first, and then progressively the next set of outer elements, the time delay between the energizing pulses being such so as to generate a wave that focuses at the required distance. The beam pattern of an annular array and a linear array transducer is shown in Figure 1b. The annular array has a symmetrical beam pattern, whereas the linear array has a narrow beam pattern similar to the annular array in the plane where the beam is generated by focusing the output of the many narrow elements, and similar to the single disc transducer in the other plane where the beam is formed primarily from the output generated by the long side of the elements.

Phased array transducers use even thinner, typically half wavelength long, rectangular elements. These transducers are used in

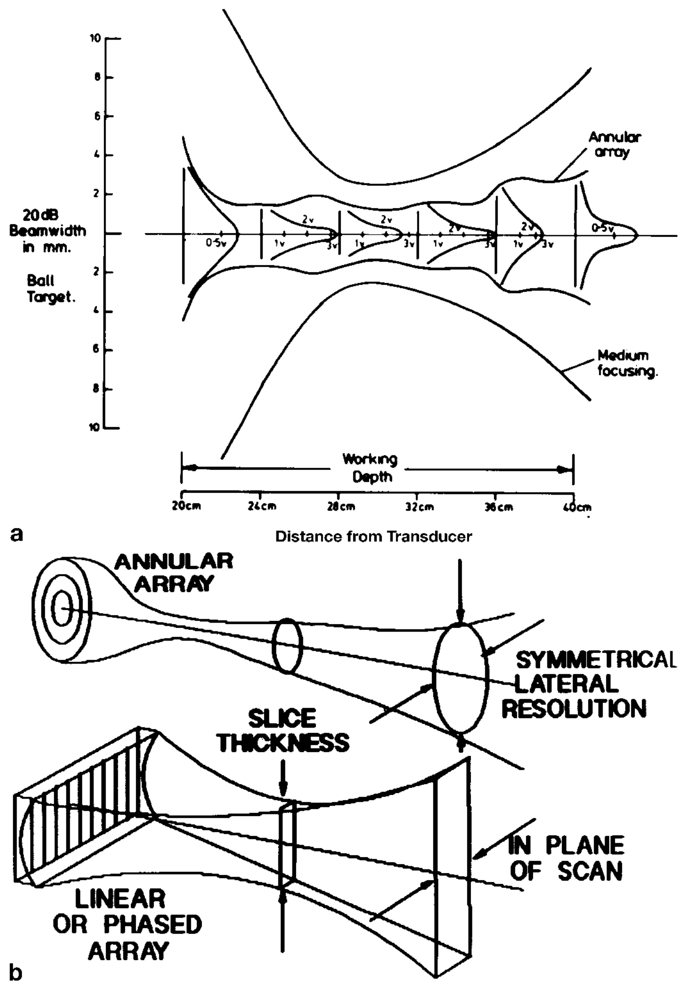

Fig. 1. a. Beam pattern of a single disc and an annular array transducer. A spherical lens has been used with the single disc transducer to provide medium focusing. The beam is narrow at the focus but diverges rapidly outside the focal zone. The beam of the annular array transducer is relatively uniform over the working depth. **b.** Beam pattern of an annular array and a linear array transducer. The annular array transducer has an axially symmetric beam pattern. The beam pattern of the linear array transducer is narrow in the length plane but is similar to that of a single disc transducer in the height plane.

applications in which the ultrasonic window or foot print is small and it is desirable to be able to steer the beam electronically. Steering is achieved by energizing the outer element at one edge of the transducer and then, with linear time delay, progressively energizing all of the elements so that the beam is steered in the desired direction. The time delays used for focusing and steering can be combined to generate a focused as well as a steered beam.

Ultrasonic Examination Modes

Diagnostic US examinations are generally performed in 3 different modes: B-mode or gray scale imaging, Doppler spectral analysis, and color Doppler velocity and amplitude imaging. Although it is possible to purchase equipment that is capable of performing only 1 of each of these examining modes, modern equipment is generally capable of performing all 3 of these modes and it is not unusual to obtain the information obtained from these 3 modes concurrently and to display it on the same image.

B-mode Imaging

B-mode imaging remains the mainstay of ultrasonic examination procedures. In this technique the transducer is energized by a short pulse and echoes from acoustic impedance discontinuities in tissues are displayed on a trace that represents the line of sight of the transducer. A cross-sectional image is formed by scanning the ultrasonic beam by either mechanical or electronic means and making the trace follow in the same scan pattern.

A major advantage of US is that these images may be formed rapidly so that they can be viewed in real time. This allows the study of the dynamics of tissue and its response to physical manipulation such as the sliding of tissues along facial planes in response to compression.

As described, in transmission it is only possible to focus the beam at 1 distance per 1 energizing pulse. If it is desired to position the focus of the beam at another depth, another energizing cycle with appropriate time delays must be employed. The ability to place multiple transmit foci throughout the examination depth is a universal feature of modern equipment. The disadvantage is that the repetition rate at which the images may be formed is reduced proportionally and this may be a serious disadvantage in the examination of rapidly moving structures.

In reception, it is possible to continuously change the focus as echoes are received from deeper structures, thus always keeping the beam in focus. This mode of operation of the equipment is known as dynamic focusing and is another feature that is now universally available on modern equipment.

The ultrasonic beam pattern has a major influence on the quality of the ultrasonic images and the way tissues and structures are displayed on these images. As shown in Figure 1a, the annular array transducer has the best overall beam pattern and therefore the best imaging capability. This is illustrated by Figure 2a,b, which is a scan of a phantom containing a number of small cystic structures throughout. The image obtained with the annular array transducer shown in Figure 2a gives an accurate representation of these cysts. Figure 2b was obtained with a high-quality curved array transducer. The cysts are portrayed accurately in the focal zone. However, despite the use of multiple transmit foci and dynamic focusing, the cysts lying close and far from the transducer are either poorly displayed or not imaged at all. This is due to the wide nature of the beam in the out-of-scan plane shown in Figure 1b which does not allow the resolution of these small structures at those distances.

The importance of accurately positioning the structures of interest in the focal zone is illustrated in Figure 3a,b. Figure 3a is a scan of a breast containing several cysts at various depths. Figure 3b is a scan in the same plane but with an ultrasonically transparent stand-off pad placed on the skin to allow positioning of the focus at the depth of the superficial cyst. The improved display of the cyst is dramatic. The boundaries of the cyst are imaged clearly and the internal content is echo free.

Despite the beam pattern advantage, annular array transducers are not commonly used in diagnostic US. Their large physical size and relatively heavy weight are major disadvantages in clinical practice. More important it is not possible with these transducers, which have to be mechanically scanned, to rapidly change the ultrasonic line of sight across the imaged plane as required, e.g., to concurrently obtain B-mode and spectral Doppler information.

Spectral Doppler Analysis

When an ultrasonic pulse impinges on a stationary interface, the frequency content of the reflected pulse is essentially unchanged. However, if the interface is moving with respect to the transducer, the frequency content in the pulse is shifted by the Doppler effect. The Doppler frequency shift is proportional to the examination frequency, the velocity of propagation in tissue, and the velocity of the interface relative to the transducer. The latter is a major cause of complication in the interpretation of the result of Doppler examinations. Investigators have to be continually aware of the angle subtended by the ultrasonic beam relative to the moving interface. An interface moving directly toward or away from the transducer gives the maximum up or down Doppler frequency shift. An interface moving with the same velocity but at right angle to the ultrasonic beam does not in effect move either toward or away from the transducer and so does not give rise to any Doppler frequency shift. Finally, an interface moving at an angle to the beam gives rise to a Doppler shift that is reduced by the cosine of the angle subtended by the beam and the direction of movement.

Spectral Doppler information can be obtained using either a continuously energized or a pulsed technique. In the former, 1 transducer is continuously energized and a separate transducer is used to receive the reflected Doppler signal. The advantages are the relative simplicity of the equipment, the good signal-to-noise ratio, which means that low acoustic outputs may be used, and the ability to measure without ambiguity very fast velocities. The disadvantage is that it is not possible with continuous Doppler to determine the depth from which the Doppler signal is originating or the interface which is giving rise to the Doppler signal. Pulsed Doppler is similar to B-mode in that a relatively short pulse is used to energize the transducer and the same transducer is used to receive the Doppler signal. The major difference is that in B-mode the ultrasonic beam is scanned to form the image, whereas in spectral Doppler the beam is kept stationary and the signal is processed to determine the amplitude and the range of frequencies present. An adjustable range gate is used to select the depth from which the signal is received and a sample gate width is employed to allow measurement of Doppler signals originating from within the selected gate width. The disadvantage is the considerably increased complexity and thus cost of the equipment

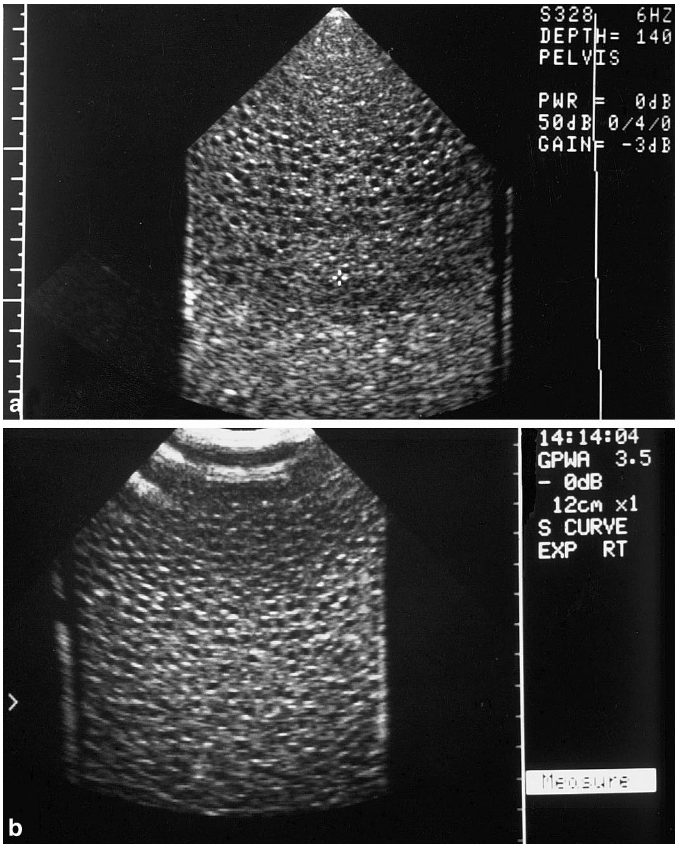

Fig. 2. a. Image of a phantom containing many small cystic structures obtained with an annular array transducer. An accurate representation of the cysts is obtained throughout the phantom. **b.** Image of the same phantom with a curved array transducer. Only the cysts lying in the focal zone are imaged clearly. The open arrowhead indicates the position of the focus.

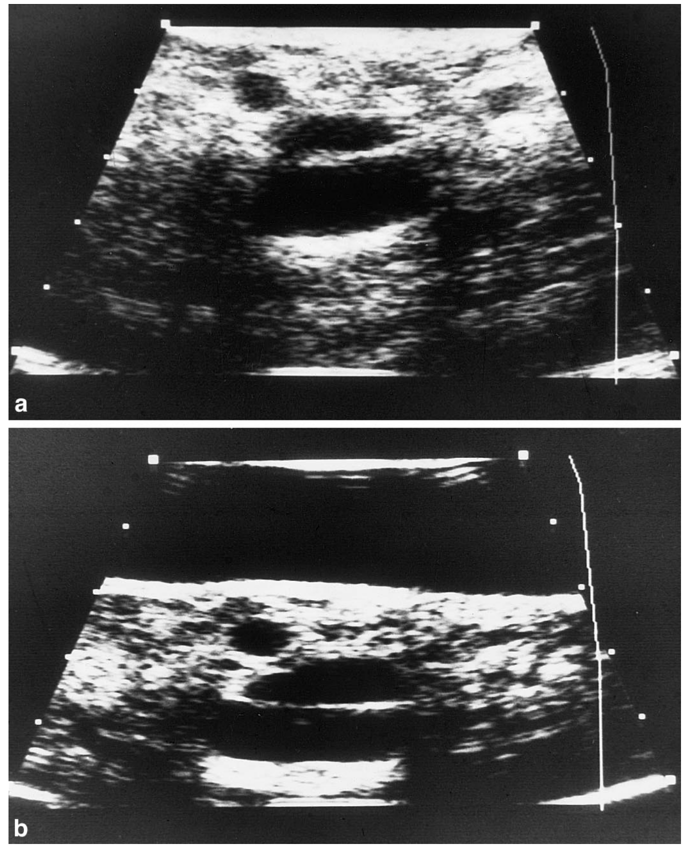

Fig. 3. a. Sonogram of a breast containing 3 cysts at varying depths. The cyst lying close to the transducer is not imaged well and could be misinterpreted as a solid mass. **b.** Sonogram of the same breast using a stand-off pad to position the cysts in the focal zone of the transducer. All 3 cysts are imaged clearly.

and susceptibility of the technique to a number of artifacts and in particular aliasing. The latter is a phenomenon that is characteristic of pulsed techniques whereby flow above a certain velocity is incorrectly displayed as flow in the opposite direction.

Spectral Doppler can be used clinically at increased level of complexity and thus level of interpreting skill. In the simplest case, a range gate can be placed over an imaged vessel to determine simply presence or absence of flow in the vessel, e.g., to distinguish bile ducts from blood vessels in the liver. In the next level of complexity, presence of steady versus pulsatile flow can be used to distinguish veins from arteries. Analysis of the pulsatile characteristics of flow requires interpretative skill and should not be attempted by untrained staff.

Figure 4a is a schematic of typical pulsatile waveforms obtained from blood flow in arteries. The top waveform is obtained in arteries with low resistance to flow. The flow increases rapidly with the onset of systole. There is then a gradual decrease to a steady but significant flow in diastole. In contrast, as illustrated in the bottom waveform, there is no flow in diastole in arteries with high impedance to flow, and in severe cases some reversal of flow immediately after systole as blood is forced back by the high resistance of the arterial vasculature and the downstream impedance.

Figure 4b illustrates 3 of the common parameters used to describe the properties of these waveforms. The systolic/diastolic (S/D) ratio is the simplest parameter whose value ranges from 1 in venous flow to infinity in high resistance arterial flow. The major disadvantage of the S/D ratio is the rapid change in the value of the ratio as the impedance to flow reduces the diastolic flow to 0.

The $(S - D)/S$ resistance index was introduced to overcome this limitation. The value of the resistance index ranges from 0 in venous flow, i.e., when $S = D$, to 1 in high impedance arterial flow where D is 0.

Neither of these indexes provides any assessment of the shape of the waveform. The pulsatility index attempts to take some account of the shape of the waveform by also measuring the mean value of the flow. This requires dedicated measurement capability of this parameter by the equipment which, unfortunately, is not universally available.

Red blood cells are the major source of interfaces that give rise to Doppler signal from flowing blood. These are physically much smaller than the ultrasonic beam and all of the cells within the ultrasonic sample gate width and beam pattern give rise to the received Doppler signal. If all of the cells flow in the same direction with the same velocity, only a single Doppler frequency shift is returned. As illustrated in Figure 5a, this situation occurs in laminar flow where the blood flow throughout the vessel is

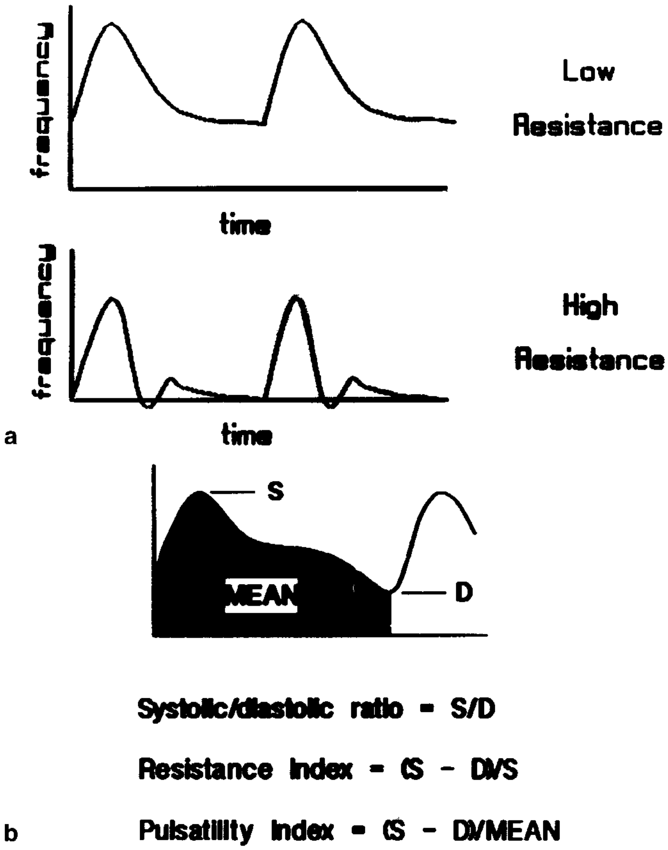

Fig. 4. a. Doppler frequency shift obtained from vessels with low and with high resistance to flow. In the case of low resistance there is continuous flow throughout the cardiac cycle. **b.** Three indexes used to describe the properties of the Doppler waveform.

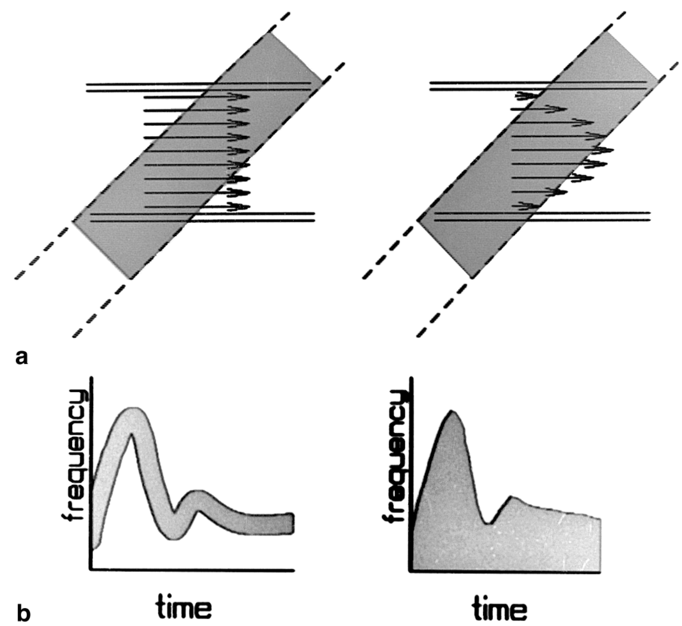

Fig. 5. a. Laminar (left) and parabolic (right) blood flow (arrows) in a vessel. **b.** Doppler spectrum obtained from laminar and parabolic flow. In the former, only a single Doppler frequency shift is obtained. In parabolic flow a spectrum of frequencies, corresponding to the different velocities present in the vessel, is measured.

uniform. Commonly, as illustrated in Figure 5b, blood flow in vessels is parabolic where the fastest flow is on axis. Red blood cells flowing with different velocities are encompassed by the sample width gate and a spectrum of Doppler frequencies is measured by the equipment. Similarly, if the flow is turbulent, a broad spectrum of frequencies is obtained. As illustrated, spectral Doppler displays the range of frequencies present and allows accurate analysis of the characteristics of flow in the examined vessel.

Color Doppler Imaging

Color Doppler imaging is in principle similar to B-mode imaging. The transducer is energized by a relatively short pulse and the Doppler information obtained along the line of sight of the transducer is displayed in color rather than in gray scale as used in B-mode imaging. A cross-sectional image is formed by scanning the line of sight of the transducer and making the trace follow the same scan pattern.

The technological requirements are, however, much more complex. The duration of the energizing pulse is a compromise between the opposing requirements of a longer pulse to allow accurate measurement of the Doppler frequencies present in the measurement pixel and of a shorter pulse to keep the pixel size small, appropriate to the desired spatial resolution of the color image. The duration of the pulse is therefore usually under computer control and is automatically adjusted as equipment settings are altered to maintain the optimum conditions at all settings.

In contrast to B-mode imaging, the line of sight of the transducer in color Doppler is kept stationary for several energizing cycles, typically 8, to allow determination of the Doppler frequencies present in the displayed pixels to the required accuracy. The color Doppler image is therefore acquired proportionately more slowly. This can lead to interpretative difficulty, e.g., in instances in which the left side of the image is acquired at one instant of the cardiac cycle and the right side of the image is acquired a short time later but in conditions in which, due to rapid change associated with the cardiac cycle, the flow has undergone significant transition. For this reason, color Doppler images are frequently acquired only over a restricted angle or portion of the B-mode image so that these images may be acquired at an appropriate time frame rate.

Calculation of the Doppler frequency shift is not a trivial task and in color Doppler these calculations are performed simultaneously on all pixels along the transducer line of sight. This massive computing task has become possible only with the development of modern digital signal processing technology.

In spectral Doppler, the frequency information is portrayed in chart form, which allows an accurate display of all of the frequencies or velocities that are present in the range gate and, by gray scale, of the amplitude or strength of the Doppler signal at those frequencies. This is not possible with a color display as 1 color can be used to display only 1 parameter of the Doppler signal.

Color Doppler velocity imaging is used to display the Doppler shift frequencies such as those due to blood flow in vessels that are present in the scanned cross section. Usually the mean of the

Table 2. Properties of color Doppler velocity imaging.

Displays magnitude and direction of flow
Measurement of velocity depends on angle of inclination
Cannot be used to measure flow at 90 degrees
Is sensitive to aliasing
Is a robust imaging technique

Table 3. Properties of color Doppler amplitude imaging.

Only indicates presence of flow
Is not dependent on inclination
Can be used to measure flow at 90 degrees
Is not subject to aliasing
Allows better imaging of noisy data
Allows better imaging of slow flow
Is sensitive to tissue motion

frequencies present in the pixel is displayed, although other options such as the display of maximum frequency are also usually available. The equipment is designed to adjust automatically for the frequency of the transducer and, when indicated by the operator, for the line of sight of the transducer relative to the flow. This allows calculation of the velocity of the flow, hence the name of the technique. It must be appreciated that the correspondence that 1 color represents 1 value of velocity only applies along the indicated line of sight. If the vessels or the flow change direction this correspondence no longer applies and this can cause interpretative difficulty for the novice. As implied by the name, color Doppler amplitude imaging displays the amplitude of the Doppler signal in the pixel. The technique is a recent development, but as it provides information that complements that obtained with velocity imaging, it is a feature that is now universally available on modern Doppler equipment. The major advantages of the technique include increased sensitivity, improved imaging of slow flow, ability to measure flow in the direction normal to the ultrasonic beam, and immunity to aliasing. The disadvantages are the increased sensitivity to tissue motion which gives rise to flash artifacts and the inability to portray the direction of the flow. The features of velocity and amplitude imaging are listed in Tables 2 and 3, respectively.

In summary, B-mode imaging is the primary ultrasonic examination technique that provides information regarding the morphology of tissue. Color Doppler complements B-mode imaging in that it provides functional information about blood flow. Velocity imaging allows rapid cross-sectional assessment of flow and identification of areas requiring closer scrutiny. Amplitude imaging complements the velocity information by allowing more sensitive imaging of flow. Color Doppler cannot be used for quantitative assessment of flow and these studies should be performed using spectral Doppler.

Imaging Characteristics of US

Ultrasonic imaging equipment is designed on the premises that the ultrasonic energy propagates through tissue in a straight line, that the ultrasonic beam is very narrow, that the acoustic properties of tissue such as velocity of propagation are uniform, and that there are no multiple reflections between tissues. In fact, the ultrasonic energy propagates through tissue as a beam of a finite size determined as described by the properties of the transducer. The velocity of propagation in different tissues varies and this can lead to the deviation of the ultrasonic beam from the assumed direction of propagation, resulting in misplotting of the position of the echoes on the image from their true spatial position. Although the acoustic impedance mismatch between soft tissues is small, this is not the case when bone and air are present in the imaged cross section. The large impedance mismatch at these surfaces can lead to the formation of multiple echoes and if these are inclined to the beam, can deviate in a mirror-like fashion the incident energy a large angle away from the original line of sight. These assumptions as well as the breakdown in other assumptions lead to the creation of artifacts that must be appreciated in the interpretation of ultrasonic images.

Enhancement and Shadowing

Soft tissues attenuate the ultrasonic energy at an average rate of 0.6 dB/cm/MHz on a "there and return" basis. This means, e.g., if a 5 MHz transducer is used for the examination and the penetration depth is 10 cm, the amount of energy that reaches a transducer from that depth is reduced by 30 dB. The amplitude of the echo is proportional to the square of the energy so that amplitude of the echo from that depth is reduced by 60 dB. Referring to Table 1, this means that the echo from that depth is one thousandth that received from superficially lying tissues.

A time gain compensation (TGC) control is available on the equipment to compensate for this loss in echo strength by progressively increasing the gain of the equipment as deeper echoes are received. The TGC control compensates for the average attenuation by tissues in the imaged section. However, it cannot compensate for a large local tissue variation and this can give rise to either an enhanced or a shadowed display of interfaces lying behind those tissues.

The liquid content of cysts does not significantly attenuate the ultrasonic energy and cysts are classic examples of structures that give rise to an enhanced display of posterior tissues. This artifactual display is, however, clinically valuable, and the enhanced display of posterior structures is an important diagnostic criterion that complements the internal echo-free appearance of cysts and confirms the diagnosis. It should be appreciated that other low attenuating tissues can also give rise to enhancement and this feature should not be taken to be pathognomonic of cysts. Shadowing is the effect opposite to enhancement which is seen behind tissues that strongly attenuate US. Tissues that are rich in fibrous content are examples of such tissues, and this is one of the reasons why shadowing is frequently observed behind malignancies.

Other effects that result in less energy reaching posterior tissues also give rise to shadowing. These include strong reflection of the incident energy by an interface due to calcification as in the case of gallstones, by bony structures, or due to presence of air. More subtle causes include broadening of the ultrasonic beam by edges of lesions such as cysts, which is referred to as diffraction edge shadowing.

Whether shadowing is obtained behind a strong reflector such as a calcification also depends on the physical size of the reflector relative to the ultrasonic beam. Macrocalcifications are generally comparable in size to the ultrasonic beam. All of the energy in the beam is therefore reflected by the macrocalcification and it will cast a shadow. Microcalcifications, on the other hand, are smaller than the ultrasonic beam. Only that component of the energy which strikes the microcalcification is reflected while the rest bypasses the microcalcification and penetrates deeper into the tissues. Microcalcifications are therefore usually displayed as small, relatively strong reflectors that do not cast shadows.

Side Lobes

The ultrasonic beam propagates through tissue as a beam of a finite size as determined by the transducer. The equipment, however, considers that the beam is very thin and echoes are displayed on the trace of the screen representing the line of sight of the transducer.

Consider a point interface lying on the axis of the ultrasonic beam as it is being scanned by a linear array transducer. In this instance the interface is displayed correctly in its spatial position. Now consider the next transmit pulse where the line of sight has been moved laterally a small distance to its next spatial position on the screen. Because, as in most equipment, the width of the beam is greater than the translated distance, the interface still lies within the beam and gives rise to an echo that is displayed incorrectly as originating from the second line of sight as well. This incorrect display continues while the point target remains within the ultrasonic beam width, resulting in the point target being displayed as a line of length equal to the beam width. This determines the lateral resolution of the equipment. Because the energy in the beam decreases toward the edges, the amplitude of the echo also decreases at the edges, displaying the target by a characteristic whisker-like appearance.

The strength of the echo from a target depends on the degree of acoustic impedance mismatch between the target and the surrounding tissues. The equipment is generally set up to display the small impedance mismatches between soft tissues. When a strong mismatch is present as in the case of bone or air, a strong echo is obtained. This strong reflector will continue to give rise to a displayable echo even when the line of sight has been moved a large distance away from the axial line of sight. Because the distance from the transducer to the reflector remains constant while the line of sight moves laterally, the reflector is portrayed as a long, spherically curved echo, the radius of curvature corresponding to the distance between the reflector and the transducer. This curved appearance of a strong reflector is commonly referred to as a side lobe. Side lobes are generally easily recognized when they originate from structures lying in the plane of the scan. They can also originate from strong reflectors lying above or below the plane of the scan. It is then more difficult to identify the anatomic origin of the echo from a single scan and unambiguous interpretation requires the acquisition of a number of scans in differing planes.

Multiple Reflection Echoes

Multiple reflection artifacts originate when the ultrasonic energy undergoes multiple reverberations in the same tissue layer. The ultrasonic transducer represents a major impedance mismatch to soft tissue, and multiple reflection echoes are usually displayed under the transducer due to multiple reflections between the transducer and the subcutaneous fat and the muscle fascial layers. Superficially lying bony structures are also a frequent source of multiple reflections. The classic example of such bony structures is the fetal skull when it is located close to the maternal abdominal wall.

Reverberation echoes are another source of multiple reflections echoes. Small air bubbles and metal foreign objects resonate when insonated by the ultrasonic pulse. These structures resonate like a bell in response to the incident pulse and give rise to a characteristic long comet-like reverberation echo.

Misregistration

The ultrasonic equipment is designed on the assumption that the velocity of propagation is constant at 1540 m/s, which is the velocity of propagation in normal saline at body temperature. In fact, the velocity of propagation varies from 1450 m/s in fat to an average of 1570 m/s in other soft tissues. This velocity variation causes misregistration of the interfaces from their correct anatomic position. It is generally customary to ignore this misregistration unless precise measurements are required. In some instances this misregistration can be utilized clinically, such as noting the forward movement of interfaces behind a lesion that can then be diagnosed reliably to be of fatty origin.

Mirror-like deviation by large strong reflectors can pose greater interpretative difficulty. For example, the diaphragm acts as a mirror reflector and can give rise to a display of lesions appearing to originate behind the diaphragm which in reality lie in the liver. These can be fairly readily appreciated when the lesion lies in the plane of the scan, but can be more difficult when the reflection is away from the plane of the scan and the lesion lies above or below the scanned plane. Velocity variations between tissues can also cause significant deviation of the ultrasonic beam and even give rise to multiple display of the same structure. The curved lens-like structure of the abdominal rectus muscle is a reflector of this type and can give rise to a double and an even triple portrayal of the same structure, transforming a single gestation sac into the appearance of a twin or even a triplet pregnancy.

In summary, propagation through tissue incurs the generation of a variety of artifacts and gradual degradation in the imaging properties of the equipment. Best images are obtained by positioning the tissues as close as possible to the transducer. This allows the selection of highest frequency and thus best axial and lateral resolution as well as best tissue contrast resolution. The clearest images are obtained with intraoperative scanning when all of the overlying tissues are absent. The next clearest images are obtained with internal scanning such as transvaginal, transrectal, and transesophageal scanning in which the deleterious effect of propagation through the subcutaneous tissues is avoided. Unfortunately, the noninvasive nature of transcutaneous scanning is mitigated by the poorest imaging performance.

Golden Rules of Sonography

As described in the previous section, the acquisition of ultrasonic images is based on a number of assumptions regarding the acoustic properties of tissue and the propagating characteristics of the ultrasonic beam in those tissues. None of these assumptions holds on close examination and this breakdown can lead to incomplete representation of the scanned tissues as well as the creation of artifacts, all of which must be appreciated in the correct interpretation of ultrasonic images. As with other imaging techniques in medicine, appropriate training is a prerequisite to the development of professional interpretative expertise. Skilled interpreters of ultrasonic images do, however, follow 3 golden rules:

1. Never make an interpretation on a single image. If a displayed feature is real, one must be able to image it in different scanning planes and from different approaches. The interpreter must learn to mentally superimpose the ultrasonic cross-sectional images to formulate a 3-dimensional image of the scanned tissues and to ensure that the displayed feature is consistent with this 3-dimensional image.
2. Because a feature is displayed it is not necessarily real. Ultrasonic images contain a variety of artifacts and it may be necessary to use a number of different scanning approaches and to apply appropriate interpretative criteria to ensure that an unusual feature on an image is not an artifact.
3. Because a feature is not displayed it is not necessarily not there. The ultrasonic beam may be deviated by tissues so that it does not follow the displayed line of sight and therefore does not display features in that area. Large inclined surfaces reflect the ultrasonic energy in a mirror-like fashion away from the transducer and are not visualized. Absence of a feature on an image therefore does not necessarily mean that this feature is not present in the section.

Conclusions

The characteristics of ultrasonic images are determined by the acoustic properties of tissues and the premises made about the propagating properties of US in those tissues. As described, sonograms provide an incomplete representation of the examined tissues and can also contain artifactual detail. Skilled interpretation of ultrasonic images is facilitated by good appreciation of the physical and technical issues that are involved in forming these images.

Résumé

Les caractéristiques d'imagerie de l'échographie diagnostique sont déterminées par les propriétés ultrasoniques des tissus. Les paramètres les plus importants à prendre en compte sont la vitesse de propagation et l'atténuation des ondes d'ultrasons. Ces paramètres déterminent la fréquence avec laquelle les tissus peuvent être visualisés qui, à leur tour, fixe la limite fondamentale de la résolution axiale et latérale. Les appareils d'imagerie ultrasonique sont basés sur le principe que l'énergie ultrasonique se propage à travers les tissus en ligne droite et que le faisceau ultrasonique est très étroit. En fait, l'énergie se propage comme un faisceau de dimensions finies, déterminées par les caractéristiques physiques du transducteur, et la façon dont celui-ci a été construit ainsi que par la forme d'énergie qui l'alimente. Aussi, la vitesse de propagation dans les différents tissus varie et ceci peut dévier le faisceau de sa direction originale. Cette déviation crée des artéfacts qui doivent être pris en compte dans l'interprétation des images. Pour ces raisons, on doit interpréter les images par des personnes expérimentées et l'interprétation doit passer par trois règles d'or: ne jamais interpréter une image isolée; ne pas considérer qu'une image correspond toujours à quelque chose de concret, et ne pas considérer que toute structure manquante sur l'image l'est également en réalité.

Resumen

Las características imagenológicas de la ultrasonografía diagnóstica son determinadas por las propiedades del tejido, siendo la velocidad de propagación del ultrasonido y la atenuación los dos parámetros más importantes. Estos parámetros determinan la frecuencia con la cual los tejidos pueden ser examinados, los que a su vez marcan el límite fundamental de la resolución axial y lateral. El equipo de imagenología ultrasónica es diseñado sobre la premisa de que la energía ultrasónica se propaga a través del tejido en línea recta y que el haz de ultrasonido es muy delgado. En efecto, la energía ultrasónica se propaga por el tejido como un haz de dimensiones finitas que establecen las dimensiones físicas del transductor, la manera como está construido y la forma como se le provee energía. Así mismo, la velocidad de propagación en los diferentes tejidos es variable, lo cual lleva a la desviación del haz ultrasónico de la dirección presumible de propagación. Tal ruptura de la presunción resulta en la creación de artefactos, los cuales deben ser tenidos en cuenta durante la interpretación de las imágenes ultrasónicas, razón por la cual, los expertos en su interpretación, observan tres reglas de oro: nunca hacer la interpretación sobre una sola imagen; el hecho de que una imagen aparezca, no quiere decir que sea necesariamente real; y el hecho de que algo no aparezca, no significa necesariamente que no esté allí.

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