ORIGINAL RESEARCH PAPER

Medical Physics

PRINCIPLES OF ULTRASOUND TECHNOLOGY IN MEDICAL IMAGING OF NON IONIZATION IMAGES

KEY WORDS: Education, X-ray tube, X-rays, Non Ionization Radiation , X- ray Radiologists & Radioprotection Protection, Medical Instruments

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ABSTRACT

The X-ray tube is one of the most important components in any X-ray system. In the beginning, physicists and physicians used gas ion tubes. The so-called Coolidge tube applied a high vacuum and is still used today. Medical examinations have required continuously improved designs of X-ray tubes (smaller focal spots at a higher output). The principle of the Goetze line focus is still applied in any diagnostic X-ray tube. Different anode materials and the rotating anode contributed to an increased output and reduced exposure time. Bearings needed special attention. Spiral groove bearings are the most advanced design today. The heat storage capacity of the anode and the tube housing assembly influences examination time and patient throughput. Cardiac imaging required less motion blurring in cine film images and increasing radiation exposure in interventional procedures called for measures to reduce dose. Protection against radiation and electric shock has always been a concern of design engineers. Focal spot sizes dedicated to specific applications and heat management within the total tube housing assembly will be future issues. Even with the event of ultrasound and MR technology, X-ray procedures will still be applied for diagnostic and interventional purposes.

INTRODUCTION

Ultrasound, like X-rays, are waves that carry energy to space. A wave is a change in some quantities, known as wave variables, that propagates in space at a characteristic speed. In the case of X-rays, which carry electromagnetic energy into the vacuum, the propagation speed is the known speed of light (c = 3 x 108 m / sec). Ultrasound, which is nothing more than sound waves with frequencies higher than those to which the human ear is sensitive (frequencies greater than 20 kHz), differs from electromagnetic waves in that it requires a means of propagation to transmit energy to space.

Typical ultrasonic velocities in various biological and other materials are given in the table

Table 1.

Material	Speed (m / sec)
Fat	1450
Air	331
Oil (castor)	1500
Water (50 ° C)	1540
Soft tissues	1540
Liver	1550
Blood	1570
Muscle	1585
Bones (skull)	4080
Quartz	5740
Aluminum	6400

Another key difference between sound and different forms of electromagnetic radiation is that sound waves are longitudinal, while electromagnetic waves are transverse. Longitudinal are called waves, in which the changes of the wave variables are in the same direction as the direction of wave propagation. Instead, they are called transverse waves, in which the wave variables change perpendicularly in the direction of their propagation. Thus, in electromagnetic waves, the intensity of the electric and magnetic fields changes perpendicularly in the direction of their propagation. Another well-known example of transverse waves is the sea waves. In contrast, sound waves propagate by placing the particles of the propagating medium in a pulsating motion around their equilibrium position and in the direction of wave propagation. Continuous (non-pulsed) is called a long wave, whose wave variables are sinusoidal functions of time. This periodic change in the value of a wave variable, starting from its value in a resting state, reaching a maximum value, passing the average calm value, descending to a minimum value and returning again to the average calm value, is repeated continuously and each repetition is called a cycle. When the transmitter stops emitting this continuous wave, the propagating medium particles return to their original resting position. Ultrasound, like all waves, is characterized by certain parameters. These parameters are frequency, period, wavelength, propagation speed, wavelength and intensity. Frequency, period, amplitude and intensity depend on the ultrasound source, velocity is characteristic of the propagation medium and wavelength depends on both the ultrasound source and the propagation medium.

Frequency (f) is the number of cycles of a wave variable per sec and is expressed in Hertz (Hz) (1 Hertz = 1 cycle / sec) or Megahertz (MHz) (1 MHz = 1,000,000 Hz).

Period (T) is the duration of a cycle and is equal to the inverse of the frequency:

$T = 1 / f$ (1)

Wavelength (λ) is the length, in space, occupied by a wave cycle. That is, the distance, in the direction of wave propagation, between adjacent particles of the propagating medium having the same displacement amplitude from the resting position. The wavelength is equal to the transfer velocity for the frequency:

$\lambda = c / f$ (2)

Ultrasound Production And Detection

The operation of ultrasonic production and detection systems is based on an instrument called an energy transducer. The use and operation of energy converters is analogous to that of speakers and microphones in the case of common sounds that are audible to humans. When the dual role of transmitters and ultrasonic receivers at the same time needs to be emphasized, energy converters are also called transceivers. Another term often used to describe an energy converter that produces sound waves is the term sound transmitter. In this book all three above terms are used alternatively, depending on the general content of the chapter which refers to these instruments.

Power Converters

An energy converter is an instrument that converts one form of energy into another. In ultrasound, this instrument is the point of contact of the entire imaging system with the patient and, thus, the diagnostic quality of the final image largely depends on its proper functioning. Also, because the energy converter performs speaker and microphone duties for the production of the ultrasound beam and for the detection of reflected sounds (reflections) respectively, its role in ultrasound systems is doubly important. In the production of ultrasound, electrical energy is converted into mechanical energy, while in the detection of reflections, the mechanical energy is converted, which they transfer to electrical energy, for the subsequent electronic processing, which is required for the construction and recording of the final image. The basic unit of the inverter is the crystal, made of special natural or synthetic materials, such as quartz and titanium lead, which present the piezoelectric effect. That is, these crystals, when receiving a short electric pulse, produce a mechanical vibration at a characteristic resonant frequency, which depends mainly on the thickness of the crystal. This vibration propagates in the soft tissues of the human body like an ultrasound beam. Conversely, when the reflections strike the surface of the piezoelectric crystal, it converts the acoustic energy (pressure) they transmit to it into electrical energy. Thus, the same crystal serves as a transmitter and a receiver. The purpose of the absorbent material used in the inverter, behind the crystal, is to shorten the vibration duration of the crystal during the ultrasonic beam generation phase, so that it can receive the reflections immediately. Without the use of absorbent material, the vibration or propagation time would be longer than the time of even the most distant reflections (reflected sounds coming from the deeper layers of soft tissues) thus preventing their proper recording and processing. Although the frequency of the ultrasonic beam produced by each inverter is the same as the frequency of the alternating voltage received as an input signal, the efficiency of the inverter and the intensity of the ultrasound are maximum when this frequency is equal to the characteristic crystal frequency.

In medical ultrasound, the crystals usually receive short-lived electrical pulses and produce ultrasonic pulses in the form of damped sinusoidal waves of medium frequency approximately equal to the characteristic frequency of the crystal. For reasons explained below, the duration and therefore the spatial pulse length of the ultrasound pulse determines the ability of the ultrasound scanner to distinguish adjacent reflective surfaces in the direction of ultrasound propagation (axial resolution). The spatial length of the ultrasonic pulses is equal to the number of cycles per pulse on the wavelength and is therefore controlled by the absorbent material (fewer cycles per pulse) and the frequency of the ultrasounds (the wavelength decreases as the frequency increases). Each natural or synthetic crystal and each ultrasound transducer is characterized by the so-called factor Q, which is indicative of the "quality" of the ultrasound it produces and the damping time of the ultrasound beam after an instantaneous electrical stimulus. The factor Q is defined based on the frequency range of the ultrasound beam as follows:

$Q = fo / f2 - f1$ (3)

where fo = characteristic resonance frequency of the crystal, f1 = frequency less than fo with intensity reduced by 50% relative to the intensity of fo and f2 = frequency greater than fo with intensity reduced by 50% relative to the intensity of fo. The above definition of Q is based on the assumption that the frequency response of the crystal or energy converter is approximately symmetric with respect to the resonant frequency. In ultrasound, transducers with small Q are preferred, because on the one hand they produce pulses with short spatial length, thus increasing the axial resolution of the method, and on the other hand they detect, without forming themselves, a wider range of reflection frequencies. The latter is due to the fact that these inverters respond to a wide range of frequencies in a stable way and thus faithfully reproduce the reflections that can be analyzed at such frequencies. In contrast, Doppler-based ultrasound methods often use high Q inverters.

MATERIALS & METHODS

Ultrasound Beam Characteristics

The surface of a piezoelectric crystal can be thought of as consisting of many elementary crystals, which pulsate simultaneously when they receive an electrical pulse as a stimulus from the system. The set of acoustic waves produced by these elemental crystals composes a continuous acoustic wave or acoustic pulse (Huygen principle), which propagates through a medium forming an acoustic beam, i.e. an ultrasound beam. The intensity of the beam at a particular point in space depends on whether at that point the individual waves contribute with or without a phase difference, which in turn depends on the relative distance of the elementary crystals from that point. This is reason, for which the intensity of an ultrasound beam is neither homogeneous nor decreases monotonously as a function of the distance from the surface of the transducer. In particular, an ultrasonic beam of an unfocused transmitter can be considered to consist of two bands with special characteristics. The part of the beam contained between the flat surface of the crystal and a certain distance (near field) is called the near band or Fresnel band and is characterized by an approximately cylindrical shape (the diameter of the beam decreases slightly as the distance from the transmitter increases) and active fluctuations in intensity. The rest of the beam, beyond the near field distance, is called the far band or Fraunhofer band and is characterized by an approximately linear increase in beam diameter with distance and a monotonous decrease as a function of both the distance from the transmitter surface and the vertical distance from the beam. center axis of the beam. Fresnel band fluctuations affect the relative intensity of the reflections, and the imaging of the tissues within this distance from the transmitter is not representative of their true composition.

The diameter of the ultrasound beam varies as a function of the distance from the energy converter and depends on its diameter and the frequency of the ultrasound it emits. The main beam of ultrasound is often accompanied by lateral lobes of intensity or energy; that is, the energy converter also produces secondary beams, which carry acoustic energy in lateral directions. Usually, at a first glance, the basic principles of ultrasound do not take into account these secondary bundles. Thus, in this section, the characteristics of the main beam only, which is directed perpendicular to the flat surface of the energy converter, are examined. In ultrasound, it is desirable to check the diameter of the ultrasound beam at various distances from the surface of the energy converter, because this determines the ability of the method to distinguish adjacent reflectors in a direction perpendicular to the direction of ultrasound propagation (lateral resolution). While this is somewhat possible with the choice of transmitter diameter and frequency, the lateral resolution that can be achieved with flat surface energy converters is always limited. The lateral resolution of the ultrasound can only be improved by focusing the ultrasound beam at a clinically desired distance from the transducer. Focusing is achieved by using focused hollow crystals, acoustic lenses or phased arrays. In these cases, the diameter of the ultrasound beam decreases within a focal band and increases beyond it. Focal length is the distance of the center of the focal zone from the transmitter and is another characteristic display parameter that must be selected correctly for each clinical application. Most energy converters used in clinical ultrasound are focused.

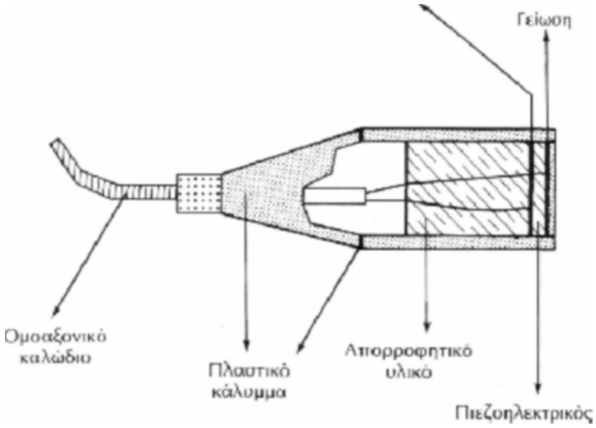

Fig 1. Ultrasonic Converter.

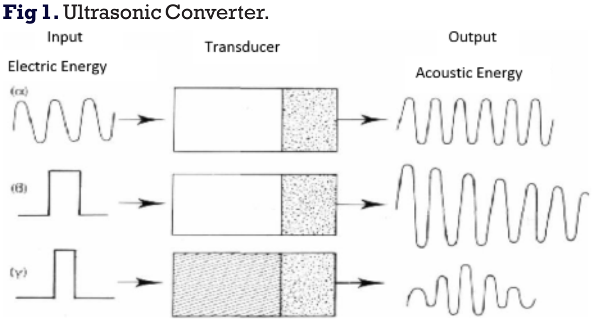

Fig 2. Typical Ultrasonic Converter Input-Output Signals

(a) Input: alternating voltage, output: continuous wave of ultrasound of the same frequency (b) Input: short electric pulse, converter without absorbent material, output: ultrasonic pulse with the characteristic frequency of the crystal and long duration. (c) Input: short electric pulse, output: ultrasonic pulse of short duration due to use of absorbent material.

Interactions Of Ultrasound And Biological Tissues

Understanding the natural principles of energy and matter interaction is important both for the design of medical diagnostic imaging systems and for the interpretation of the diagnostic content of the images they produce. The interaction phenomena of ultrasound and interactive tissues, which affect the ultrasound and examined below, are the reflection, refraction and absorption of the ultrasound beam. Unlike radiographic methods, where the imaging of the anatomy is based on the partial absorption of X-rays by biological tissues, in ultrasound the ability to image the anatomy is based on the phenomenon of reflection. The refraction and absorption phenomena of the ultrasound beam mainly have a negative effect on the ultrasound methods, affecting the relative intensity of the reflections and the accuracy with which the elementary volumes of biological tissues from which these reflections originate are located in space. Thus, in addition to errors in the relative intensity of different areas of an ultrasound, technical errors of a geometric nature can also occur, which often distort the shape of various anatomical organs or result in their misplacement. This is the reason why the correct clinical application of ultrasound requires technical knowledge, perhaps more than other methods of diagnostic imaging.

In ultrasound, the construction of the image is based on the partial reflection of the ultrasound beam, as it successively strikes at separating surfaces between tissues with different specific acoustic resistance, as defined below. The percentage of energy that is not reflected, but penetrates the various layers of biological tissues of the human body or is absorbed by them, does not serve directly in the construction of the ultrasound; however, some of this energy is then reflected from other layers of tissue allowing imaging. of anatomy to a greater depth. It is obvious that it is not possible to take an image of the anatomy beyond some distance from the ultrasound transmitter, if it has been preceded by total reflection of the beam or if it, due to the gradual absorption of its energy by the propagating medium, is not strong enough.

Simultaneously with the development of computed tomography methods with X-ray beams, computed tomography methods from anatomy projections with ultrasound beams began to develop, but to date have not yielded satisfactory results. Thus, this chapter does not examine methods of mathematical reconstruction of ultrasound scans from projections, nor methods of acoustic holography, which are also under research and have not yet found direct clinical application.

On the surfaces separating two propagation media with different specific acoustic resistance (G), partial to almost total reflection of the ultrasound beam is observed. The specific acoustic resistance is an elementary property of matter and is given by the product of density over the speed of sound in that material:

$Z = \rho * c$ (4)

where ρ is the density in g / cm3 and c is the speed of sound in cm / sec. Thus, the specific acoustic resistance of different means of propagation is given in units g / (cm2 • sec) while, if the values of acoustic resistance are multiplied by 10-5, the specific acoustic resistance is expressed in Rayls, ie 1 Rayl = 1 g / (cm2 • sec) x 10-5. Typical values of acoustic resistance of various biological tissues, piezoelectric crystals and other materials are given in the table.

Table 2.

Material	Z (Rayls)
Air	0,0004
Fat	1,38
Water (50°C)	1,54
Brain	1,58
Blood	1,61
Kidney	1,62
Liver	1,65
Muscles	1,70
Eye Lens	1,84
Piezoelectric Polymers Materials	4,0
Head (Scalp)	7,8
Quartz	15,2
Mercury	19,7
PZT-5A	29,3
PZT-4	30,00
Bronze	38,0

In addition to the vertical impact of an ultrasound beam on a surface, angular impact can also occur. Then, phenomena similar to those of optics are observed. Specifically, as in the case of vertical impact, part (of the energy) of the beam is reflected and the rest continues to propagate, but not in the same direction as the original beam. The ultrasound beam, which is formed behind the reflective surface, shows the phenomenon of refraction, ie it changes direction with respect to the original beam, due to the different speed of sound in the two means of dissemination. Ultrasound beam refraction is the main source of technical errors of a geometric nature, which often distort the shape of anatomical organs or display them in the wrong position. As explained below, the ultrasound scanner is deceived by refraction, just as the human eye is deceived when it sees a straight piece of wood, partially submerged in water, as if bent.

The above geometric view of the phenomena of reflection and refraction presupposes that the wavelength of ultrasound is very small in relation to the dimensions of the reflective surfaces. Reflections of this type are called specular reflections and look like the reflection of a beam of light from the surface of a mirror. However, if the dimensions of the reflecting surfaces and the irregularities they possibly have are comparable to the wavelength of the ultrasound, then scattering of the ultrasound beam is observed, as in the case of a light beam that diffuses as it passes through from fog and restricts our vision. However, in ultrasound, the scattering of the ultrasound beam is very important because, in addition to the imaging of large dividing surfaces of anatomical organs allowed by mirror reflections, it enables the imaging of the parenchyma of the various organs, thus obtaining useful diagnostic information. In particular, the intensity of that part of the beam, which is scattered backwards (backscatter) by uneven surfaces or heterogeneous propagation media, depends more on the frequency of ultrasound and the characteristics of the scatterers, while it is relatively independent of the angle of incidence. The scattering of the beam, and especially the scattering of part of the beam backwards, ie towards the transceiver (rear scattering), allows the display and characterization of tissues with some scattering distribution and even better imaging of reflective surfaces surrounding such tissues and it happens not be perpendicular to the ultrasound beam, so as to give strong mirror reflections.

Due to the large number of scatterers that the ultrasound beam encounters as it propagates through biological tissues, it is possible for many reflections to reach the energy converter together in such a way as to either add to the total signal output or override and produce a patient. signal. The result of this phenomenon is that the appearance of the various tissues in the final image depends not only on their actual composition and the characteristics of the scatterers they contain, but also includes characteristic fluctuations in volume, known as acoustic speckle, that come from the contribution that leads to amplification or reversal of multiple reflections on the surface of the energy converter.

Mirror reflections and scattering of part of the ultrasound beam by the tissues of the human body remove energy from the beam, thus gradually reducing the amplitude and intensity of the ultrasound.

A third phenomenon that contributes to the gradual weakening of the ultrasound beam is the absorption of energy in the tissues, by the conversion of acoustic (mechanical) energy into heat. A detailed examination of the rather complex mechanisms of absorption of ultrasound energy by soft tissues goes beyond the contents of a chapter on basic principles. Thus, in the rest of this paragraph, the phenomenon of attenuation of the ultrasound beam is completely examined, due to mirror reflections, scattering and absorption.

The attenuation factor (a) represents the attenuation of the ultrasound beam per unit distance and is expressed in dB / cm.

RESULTS & DISCUSSION

In the previous we referred to methods of production and detection of ultrasound, as well as phenomena of interaction of ultrasound and biological tissues. Ultrasound is a method of diagnostic imaging based on information produced as a result of these phenomena, as transmitted to the energy converter by reflected sounds. Today, almost all ultrasound systems emit ultrasound pulses and the image is generated after processing the information contained in the composite signals, which is produced by the energy converter in response to reflected sounds, which receives after mirror reflections and rear scattering of part of the energy. The following describes the systems and methods of static and dynamic diagnostic ultrasound.

Static Imaging

A typical diagnostic ultrasound system consists of a pulser, a power converter, a receiver or processor of the output signals of an energy converter, some video memory, and one or more TV screens. The pulsator sends electrical pulses to the energy converter at a predetermined frequency. For each electrical pulse it receives, the energy converter produces a corresponding ultrasonic pulse. Successive ultrasound pulses travel through the tissues of the human body and are reflected or scattered backwards, gradually losing some of their original energy. Reflections produced by various reflective surfaces or distributed scatterers are directed to the energy converter, where they are converted into electrical signals, the whose amplitude is proportional to the amplitude or intensity of the reflections. These signals go through various stages of processing, which lead to the construction and recording of the image in memory and then to its display on the screen.

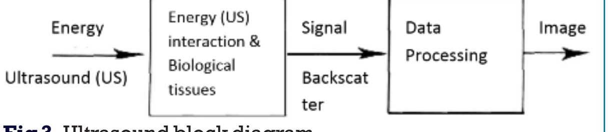

Fig 3. Ultrasound block diagram.

The processing steps of the output converter output signals are as follows:

Initially, the signals are amplified to reveal relatively low-intensity reflections and to facilitate further amplification. Adjustments are then made to the approximate amplitude of the signal to compensate for possible intensity differences between different reflections coming from reflectors with the same reflection coefficient but at different distances from the energy converter (compensation). The final image represents the reflective properties of the various organs and tissues and its intensity or brightness should not depend on the distances of the reflectors from the source of the ultrasound beam. In the next step, the signals, which initially have a frequency band of 110 MHz, are filtered or demodulated so that the envelope of the signal remains, ie the low frequencies that represent the relative magnitudes of the various reflections (demodulation). Then, due to the limited dynamic scale of the screen, in which the image is finally projected, the sizes of the various reflections are compressed in a way that ensures that the maximum difference between them is compatible with the dynamic scale of the screen (compression). Finally, reflections that do not exceed a value of amplitude or intensity are rejected (rejection). This excludes from the final image those reflections that do not contain useful information, because they are at the level of electronic or acoustic noise, e.g. multiple scattering within the tissues.

In static ultrasound, an incision of the anatomy is scanned by the ultrasound beam with a corresponding movement of the energy converter on the surface of the patient's body.

Throughout the scan, the reflections are converted into electrical signals, which go through the above processing, thus giving the information needed to compose an image, of the supposed static anatomy, directly into the system memory. In particular, these signals represent the amplitude of the reflections as a function of time, and therefore allow the calculation of the distance from the surface of the energy converter and the characterization of the reflective properties of each surface or group of dispersers. But how do one-dimensional output signals of a power converter transform into two-dimensional images? The method described below initially became known as Brightness mode, B-scan and, more recently, gray-scale imaging. This way can be easily understood, if we imagine the anatomical section divided into many cells (picture elements or pixels) and for each such cell a place in the image memory, ready to accept a value proportional to the intensity of the reflections coming from it. The distances of the reflecting surfaces from the energy converter are calculated based on the velocity of the sound in the soft tissues and the total propagation time of the initial ultrasound pulse to the point of reflection and return of the reflection to the energy converter. The table of volumes, which is thus stored in the image memory, is read and transferred to the screen with a frequency that allows the stable and continuous display of the anatomical section.

The number of cells into which the anatomical section is subdivided, and therefore the size of the memory, determine the resolution of the image. For example, when the desired display depth is 20 cm and the memory has a capacity of 512x512 elements, the resolution is of the order of 0.4 mm, while a memory of 1024x1024 elements would give a resolution of 0.2 mm. The memories that have been used from time to time in ultrasound systems are of two types, analog and digital. Analog memories use an electron beam, which stores in the memory elements an electric charge proportional to the intensity of the reflections. Modern ultrasound systems, as well as computers, use almost exclusively digital memory, in which the intensities of the reflections are stored in binary form, ie as numbers consisting of digits that take only the values O and 1. These digits are called, In the terminology of digital systems, bits and the number of bits available to store each volume, also known as memory depth, determine the minimum volume difference that can be represented in the system memory. For a certain intensity scale, the number of intensity levels, which can be represented by N bits, is 2v That is, 1 bit with values O or 1 allows the representation of only 2 different intensities, 2 bits allow the representation of 4 intensities, 3 bits the representation 8 intensities etc. Systems that store and process images typically use up to 8 bits, allowing the display of 28 = 256 intensities. These distinct intensity levels are also called quantization levels, because, in practice, the actual. intensity values are necessarily quantized at these levels. Thus, the more bits the memory has, the better the ability to distinguish small intensity differences.

The screen on which the image is finally projected also has a limited ability to distinguish many levels of intensity. The ability to distinguish multiple levels of intensity in the final image is called gray-scale or contrast resolution and is determined by the choice of memory depth in bits and screen quality. Compared to analog memories, digital memories have more stable performance and better image quality. Also, the fact that the image is stored as a numeric value table allows the image to be processed before it is displayed on the screen, in order to reduce noise, selectively amplify different areas of the image or different intensities as well as the ability to analyze of the image, in order to find and characterize the size or shape of various objects, characterize the properties of various tissues, etc.

In addition to the above way of organizing information as an image, ultrasound has traditionally used other ways of organizing information and displaying it on a cathode-ray tube or television screen. A general description of these methods is given below, mainly for historical reasons, without much emphasis on technical details. The first way is called A (A-mode) and is simply a one-dimensional graph of the amplitude of the reflections as a function of the distance from the energy converter. This representation is usually written on the screen of a cathode ray tube from an electron beam, which is directed by the electrical signal produced by the processor of the output signals of the energy converter. Specifically, for each reflection, the electron beam deviates from the horizontal scanning direction, thus writing a pulse whose amplitude is indicative of the amplitude of the reflection, while the distance of the pulse from the left edge of the screen is indicative of the distance of the respective reflecting surface. from the power converter. This way of representing is not very useful except as the first step of the second way of representing the reflections, called TM (Time-Motion mode). While mode A locates the reflective surfaces at a specific point in time, while giving the amplitude of the reflections, the TM mode, also known as the M mode (Motion mode), graphically shows the movement of the reflective surfaces along its axis. ultrasound beam. This is achieved by replacing the deviation of the electron beam of mode A by modulating its intensity according to the amplitude of the reflections (B-mode) and by using a recording medium that moves at a constant speed in the vertical direction in front of the screen. Due to its scan electron beam in the horizontal direction and the motion of the recording medium in the vertical direction, we get an image, the horizontal dimension of which is constant and equal to the maximum distance of the reflectors from the energy converter, while the vertical dimension is variable and depends on the desired time of observation of the movement of the reflective surfaces. The intensity of the image is shaped, as mentioned above, by the width of the reflections.

Dynamic Illustration

In static ultrasound, the energy converter remains stationary as it emits acoustic pulses and records the reflections (imaging modes A, M and TM) or, for anatomical imaging, moves on the patient's surface under manual control (scan mode B). In the second case, which is mainly of interest to us, the manual control of the movement of the energy converter is very slow and does not allow the dynamic depiction of the anatomy in real time (ultrasound).

In dynamic ultrasound or real-time ultrasound, as it is also known, the detection and processing of reflections, the storage of their width in the appropriate memory location and the display of the final image on the screen must be done at the speed of many images per second. The standard speeds required to give the feeling of the constant movement of various organs, such as the heart, are 20-60 images / sec. These frame rates require the use of other energy and memory converter technologies as well as other signal and image processing methods. However, apart from some differences in the hardware used and the addition of the imaging rate, as another parameter, the static imaging mode B and the dynamic ultrasound do not differ in the general principles of image construction and projection, mentioned above. . Dynamic ultrasound systems are also used today for static imaging of anatomy, as they have the ability to freeze images on the screen using image memory.

Of the advanced technologies used in dynamic ultrasound, the one we are most interested in is the technology of energy converter systems. The energy converter systems used in dynamic anatomy imaging are of two types: (a) mechanical scanning and (b) electronic scanning. Both types of energy converters allow various sections of the anatomy to be repeatedly scanned at high speed. Mechanical scanning systems consist of one or more energy converters, which perform periodic oscillation-type motion or rotate, thus scanning a angular section of the anatomical incision, while the whole system remains stationary on the patient's surface. In some systems, ultrasound scanning is accomplished with an oscillating acoustic mirror mounted in front of a fixed power converter.

Electronic scanning systems use linear or concentric arrays of small piezoelectric crystals (transducer arrays), having a rectangular or ring shape respectively. Linear devices have two modes of operation. In the first way, electric pulses are applied, simultaneously, to small groups of piezoelectric crystals starting from one end of the device and going to the other. These groups act like a crystal, thus producing an ultrasound beam, which is serially transferred from one group to another to scan an incision of the anatomy in the direction of the device. This method is equivalent to the manual linear transfer of the energy converter to static ultrasound, but the scanning speed is much higher due to the electronic control. Linear devices, which work in this way, are simply called linear devices and are different from linear phased arrays or simple phased arrays, whose piezoelectric crystals are individually excited by electrical pulses that, however, reach each with different delay. In this way, the acoustic waves produced by the various piezoelectric crystals of the device have small phase differences and constitute an ultrasound beam, the shape and direction of which change, as do the arrival times of the electrical pulses in the crystals. Therefore, linear phase device systems allow the electronic control of both the direction of the ultrasound beam and its focal length, which can now be changed dynamically. Scanning the anatomy with a linear phase device is equivalent to mechanically scanning an energy converter that performs periodic oscillating motion, with the additional possibility of variable focal length. Linear phase arrays can focus the beam only at the scan level. Focusing in the vertical direction in this direction requires the use of an acoustic lens. Concentric phase arrangements provide a uniform focus in space, but do not allow control of the beam direction, which is achieved by placing an acoustic mirror in oscillating motion in front of the concentric arrangement.

Technical Errors

The following description and analysis of the technical errors (artifacts) of diagnostic ultrasound is limited to errors due to acoustic phenomena and lead to substantial errors in the presentation of reflections in the final image. Other sources of error are optical illusions (errors of perception), the human factor (errors of interpretation), various types of noise, etc. The technical errors of diagnostic ultrasound can be classified into various general groups, based on the natural phenomena and imaging properties, examined in previous sections of this chapter. Many of these errors are obvious and do not pose significant problems in the diagnostic interpretation of the images. Others are less obvious and can lead to misinterpretation of the diagnostic content of the images, while some technical errors often indirectly indicate the correct interpretation. Therefore, it is important that the user of diagnostic ultrasound systems, in addition to medical knowledge and general knowledge of the basic principles of operation, has a clear understanding of possible technical errors, so that he can avoid pitfalls or use them as additional information that support the diagnostic process. The following are the main errors in the hearing or presentation of the anatomical information contained in the reflections. The description is accompanied by explanatory diagrams and corresponding ultrasound scans.

The technical errors of diagnostic ultrasound belong to different groups depending on their origin. Thus, errors in resolution, ultrasound direction, attenuation, and other origins can occur.

A. Resolution Errors

These errors result from the limited axial and lateral resolution of the ultrasound systems, as defined above, as well as from the section thickness, as defined by the beam dimension perpendicular to the scan plane. Lack of resolution results in the reflection of two reflectors as one or the misrepresentation of the size of very small reflectors, due to the fact that the minimum dimensions of objects on the screen are determined by the diameter of the ultrasound beam and half the spatial length of the ultrasound pulse. Both of these phenomena are technical errors that affect the fidelity of imaging the texture of soft tissues. It should be noted that tissue texture is a useful element in the diagnosis of many diseases. The finite thickness of the section allows reflections from layers of tissue with different acoustic properties to contribute to the energy converter and produce total intensities that do not faithfully represent the reflective properties of the tissues of the particular incision. The characteristic appearance of the tissues is a result of the above phenomena.

Direction Errors

The technical errors related to the direction of propagation are of a geometric nature and can be easily explained taking into account the real and apparent direction of the ultrasound beam in the tissues, after multiple reflections and refractions on the various reflective surfaces. The actual direction changes after each reflection or refraction (except for reflections from surfaces perpendicular to the beam), while the apparent direction is that assumed by the system based on the direction of the energy converter when it emits an acoustic pulse. This group presents reverberations, refraction, multidirectional, reflection errors, and errors due to the side lobes or grating lobes of single-crystal energy converters and linear or concentric arrangements, respectively.

Attenuation Errors

This group of technical errors is mainly caused by phenomena of artificial attenuation or enhancement of the intensity of reflections and includes the errors of shading due to fading and shading due to refraction (edge shadowing), amplification and focal enhancement. Shading errors occur in cases where some reflectors are located behind reflective surfaces with a high reflectance or tissues with a high attenuation factor. Also, shading of reflectors can be observed when they are behind the tips of objects that focus the ultrasound beam, due to the refraction of different parts of the beam at different angles. In contrast, amplification errors are observed behind reflective surfaces or tissue layers with a small reflection and attenuation coefficient, respectively, as well as behind peaks of objects focusing the ultrasound beam. Focal magnification error refers to the difference in reflection intensities from reflectors with the same reflective properties, but located at different points along the axis of the ultrasound beam. In particular, reflectors whose size is approximately equal to the diameter of the ultrasound beam at the focal length of the energy converter will give more intense reflections in the focal band than they would give if they were in the near or far band. Technical attenuation or amplification errors are shown below.

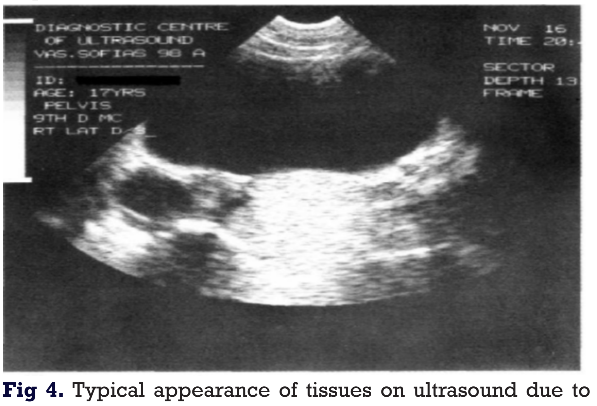

Fig 4. Typical appearance of tissues on ultrasound due to technical errors of resolution.

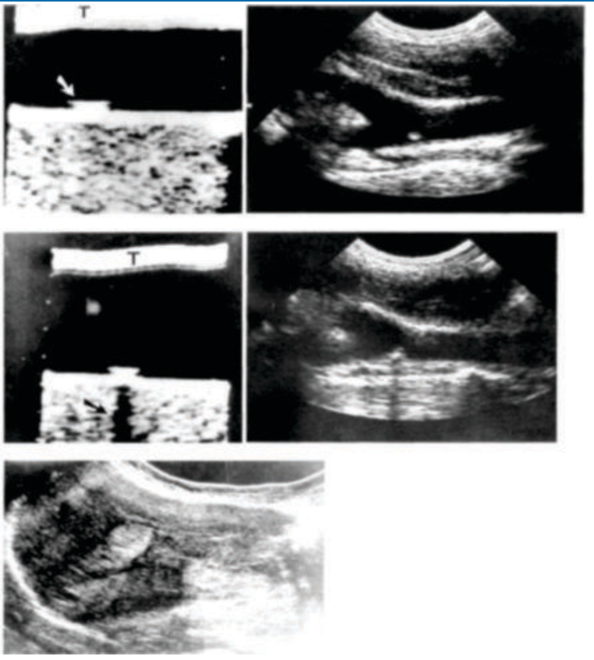

Fig 5. Technical errors of amplification or attenuation.

CONCLUSIONS

Doppler ultrasound methods are non-invasive methods for qualitative and quantitative characterization of circulatory system function. They are particularly useful in characterizing blood flow to the heart and peripheral blood vessels, and it is hoped that Doppler ultrasound will play an important role in assessing the adequacy of valves and the mobility of the walls of the heart in characterizing blood flow. in the coronary arteries and in the characterization of the elastic properties of the arteries for the early diagnosis of diseases of the circulatory system. In recent years, the clinical capabilities of Doppler ultrasound to measure hemodynamic parameters have become widely accepted. Ultrasound systems that combine the capabilities of anatomy imaging and circulatory system examination with Doppler pulsed ultrasound are already used in diagnostic medical imaging. In addition, real-time imaging systems of blood flow fields in various organs, atria, and blood vessels have recently been developed. This ability, as well as the ability to measure hemodynamic parameters at specific anatomical sites, are beginning to create additional opportunities for diagnostic use of these methods. The Doppler effect was first observed by Christian Doppler in 1843 and refers to the increase or decrease in the frequency of a continuous sound wave when it comes from a source approaching or moving away from the observer, respectively. This frequency shift is called Doppler shift and is the basis of all Doppler ultrasound systems.

For example, ambulance sirens seem to make a louder sound as they approach, while the frequency shifts to lower frequencies as they move away.

There are two types of Doppler ultrasound systems:
1. Continuous Acoustic Waves (CW Doppler) and
2. Pulsed Doppler.

Although most modern Doppler ultrasound systems use pulsed methods, the Doppler effect can be better explained by continuous acoustic waves. Continuous acoustic wave systems use two small energy converters, one as an ultrasonic transmitter and one as a receiver. The transmitter produces an acoustic wave with a frequency of 2-15 MHz, which is reflected or backscattered by moving targets, such as the walls of the heart or red blood cells, and returns to the receiver at different frequencies. Due to the fact that the target moves, as the acoustic wave falls on it, the frequency of the reflected wave differs from the initial ultrasound emission frequency and this difference is proportional to the target speed. Doppler pulsed ultrasound systems solve the problem of locating the vascular space from which the frequency shift occurs, in the same way that the various reflectors and dispersers are located on ultrasound. However, ultrasound pulses consist of a wide range of frequencies and the calculation of Doppler displacement in these systems is problematic. It is not advisable to describe in detail the signal processing methods used in this case; it is sufficient to note that doubts about the position of moving targets in space can only be reduced at the expense of the accuracy with which speed is measured. In most clinical applications of Doppler methods, the sample volume contains many moving targets (red blood cells or other scattering), moving at different speeds and in multiple directions. Therefore, in the case of CW Doppler, the ultrasonic scattering of this volume gives a complex signal consisting of many frequencies, one for each different direction and speed. In the case of pulsed ultrasound, each pulse consists of a range of frequencies and each frequency is subject to different Doppler displacements, depending on the speed and direction of the moving targets. The signal composed by these shifted frequencies is even more complex and less representative of the actual velocity distribution of the moving targets within the sampling volume. Thus, the diagnostic interpretation and general usefulness of the information provided by the clinical application of these methods depends on the techniques of processing these signals to decode the information and calculate the velocities. The ability to fully decode the information contained in the signals produced by Doppler methods is subject to basic limitations due to natural phenomena, such as the dependence of the attenuation of ultrasound on biological tissues and the scattering coefficient of the various reflectors on the frequency, which make it accurate calculation of Doppler displacements is very difficult. Many researchers today deal with the solution of problems in this area, aiming at the quantitative measurement of hemodynamic parameters by Doppler methods.

Biological Effects

These risks come from the biological effects of the amount of energy absorbed by the tissues of the human body and place restrictions on the safe use of any diagnostic imaging method. Therefore, all those involved in both the design and manufacture of diagnostic ultrasound instruments and Doppler ultrasound as well as their clinical application must be aware of the relative risks of application of these methods (risk) and always weigh them in moderation of their diagnostic utility (benefit). It is worth noting, however, that the biological effects of ultrasound and other forms of energy are the basis of many therapeutic methods (radiotherapy, hyperthermia, etc.).

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