17. Ultrasound Diagnostics

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Today, ultrasound diagnostics is an important imaging method in virtually all medical fields. The fact that it is quick, simple and in particular cost-efficient plays a major role in this. Further advantages are provided by the mobility and the broad spectrum of use of modern ultrasound diagnostic systems. Not least, these properties and also the absence of ionizing radiation make its use indispensable these days.

According to data from the German Electrical and Electronic Manufacturers' Association (Zentralverband der Elektrotechnischen Industrie – ZVEI), the ultrasound market for new devices has a sales volume in Germany of approximately € 200 million per year. Approximately 50% of this volume is from acquisitions in clinics and hospitals, and approximately 50% is from doctors in private practices. The distribution of sales between the different specialist fields gives the following picture: approximately 30% gynaecology, approximately 30% internal medicine and approximately 20% cardiology.

The high density of devices used in everyday diagnostics is due to there being a large number of different technical and applicative methods (Table 17.1) and a number of different industrial suppliers. Ultrasound diagnostics has partly replaced or supplemented other methods, such as conventional x-ray diagnostics but

Table 17.1 Overview of the advantages and disadvantages of imaging methods

	Ultrasound	CT	MRI	X-ray	Angio	Nuclear medicine
Ionizing radiation	No	Yes	No	Yes	Yes	Yes
Real-time	Yes	No	No	No	Yes	(Yes)
Type of image	Slice	Slice	Slice	Section	Section	Section
Overall costs	Low	Very high	Very high	High	High	Very high

also computer tomography (CT) and magnetic resonace imaging (MRI).

In contrast to these methods, ultrasound is a so-called real-time method. The organs being investigated are displayed on a monitor in real time in this sectional imaging technique. They therefore correspond to a tomographic sectional image familiar from CT and MRI. Ultrasound imaging thus differs substantially from the method of looking through the body in a conventional x-ray examination.

17.1 Basic Physical Principles

17.1.1 Principle

The following explanations regarding the generation of ultrasound images follow theoretical considerations and are presented in an idealized form. They are based on physical principles which are intended to be presented to the reader in a way which is simple and easy to understand. The actual generation of ultrasound images is much more complex, however. An essential part of image generation is based on the phenomenon of scattering, which makes the process which actually takes place much more complicated, and this will only be referred to here in passing. It will not be discussed in further detail in the text below.

In ultrasound diagnostics, the transmitter and receiver are combined in the ultrasound probe. The probe is connected to the ultrasound unit via a cable, thereby enabling very free positioning of the ultrasound probe on the body and thus also virtually any desired examination plane and slice orientation.

In contrast to the x-ray method, which is a transmission method, ultrasound is a so-called reflection method. The ultrasound probe transmits a short ultrasound pulse which penetrates the body and is partially reflected at interfaces, e.g. between liver and kidney. Once the pulse has been emitted, the unit switches the ultrasound probe to receive mode. The ratio of transmission to reception time is approximately 1 : 1000.

The reflected components of the sound wave transmitted are then recorded by the ultrasound probe and fed to the unit for further processing. With the exception of the continuous wave method (Sect. 17.3.2), ultrasound units therefore basically work with what is known as the pulse method. This principle is also wide-spread in the natural world and is used, for example, by bats, dolphins and other animals for orientation.

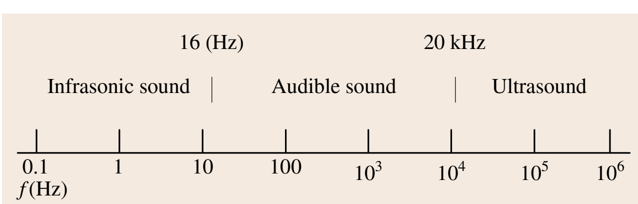

Fig. 17.1 Frequency ranges

17.1.2 Generation of Sound Waves

Special, cut piezoelectric crystals are used to generate sound waves. These synthetic crystals are manufactured industrially, for example from fractions of barium titanate, lead zirconate, lithium compounds or other ceramics. These are subject to what is known as the piezoelectric effect, which was described by the brothers Jacques and Pierre Curie using crystals of tourmaline. As soon as a voltage is applied to a crystal of this nature, it changes its form or geometry. Depending on the polarity of the voltage, dilatation or contraction of the piezoelectric crystal occurs.

If a high-frequency alternating current is therefore applied, then it oscillates at precisely that frequency and generates high-frequency sound waves. If the alternating current applied is at a correspondingly high frequency, this produces ultrasound waves which are no longer audible to humans (Fig. 17.1). In the receive mode, however, sound waves impinging on the crystal are converted into an electrical alternating current which is then processed further by the unit (Sect. 17.3).

17.1.3 Reflection

On the way through the tissue, components of the transmitted sound wave are reflected at interfaces between different organs and sections of organs. The energy of the waves which are reflected is determined in each case by the differences in the so-called wave impedance (Z) of the individual organ parts and sections scanned. There is more or less reflection at these multiple interfaces depending on the relationship between the individual wave impedances, on the angle of impact with respect to the interface and on the surface texture (scattering).

In order to contribute towards the generation of an image, the reflected wave must have sufficient energy as energy is also lost on the return path to the crystal. The ideal reflected component is approximately 1%, which corresponds to a reflection coefficient of $R = 0.01$. Higher reflection coefficients result in a greater loss of energy of the propagated wave overall, thus resulting in insufficient penetration of the ultrasonic beam into deeper tissue.

The reflection coefficient is by definition between 0 and 1, where $R = 0$ means that there is no reflection taking place and $R = 1$ corresponds to total reflection. The reflection coefficient R is calculated from the wave impedances of the two media which form the interface (Fig. 17.2). The difference between the two wave impedances is crucial here.

As can be seen from (Fig. 17.3), the wave impedance of air is vastly different from that of human tissue, with the result that there is a high reflection coefficient of nearly $R = 1$ and total reflection occurs. An interface with air is therefore an obstacle to ultrasound which cannot be overcome. In order to achieve the best possible transmission from the probe to the body without any loss of energy, ultrasound gel is used as a coupling medium. On account of its very high water content, the gel conducts sound well and should ensure air-free contact with a reflection coefficient of roughly $R = 0$.

However, it is not only air but also metal parts, bones and calcium particles which can present problems that produce virtually total reflection. This explains acoustic shadows behind bones, gall stones or endoprostheses, for example.

17.1.4 Spatial Mapping – Transit Time

The spatial mapping of the scanned tissue and reflecting interfaces is done by measuring the transit time from transmission of the pulse to receipt of the respective reflected components. Early echoes means that the waves have been reflected by interfaces close to the ultrasound probe. Late echoes means that the waves have been reflected by interfaces a long way from the ultrasound probe. This difference in time is represented on the monitor display by a corresponding distance between the pixels in the sound beam direction. However, this presupposes knowledge of the speed of sound conduction in the tissue, which is assumed to be on average 1530 m/s (Fig. 17.4). Since different tissues also have different wave impedances, this average is inevitably a generally accepted compromise and is used in standard commercial ultrasound units.

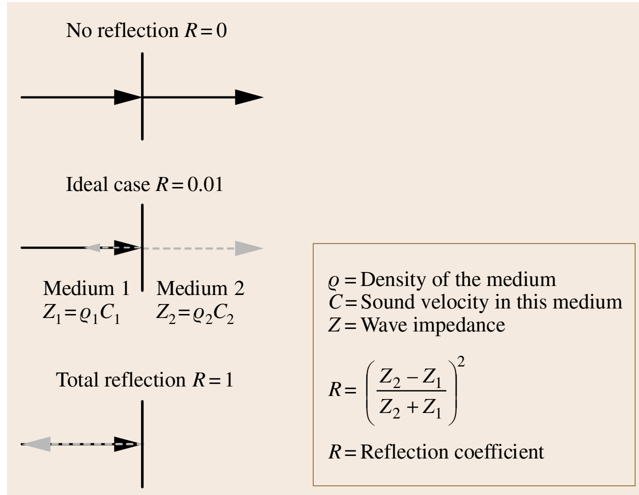

Fig. 17.2 Reflection coefficient and wave impedance

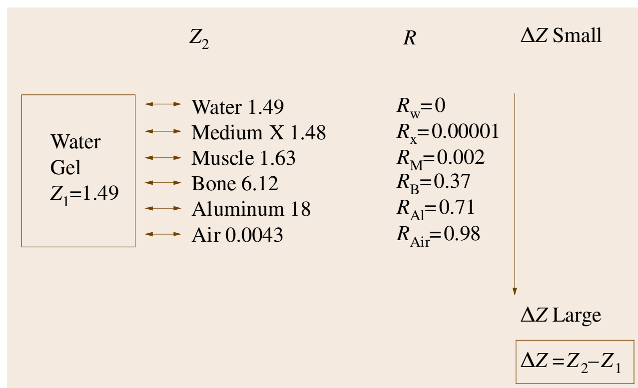

Fig. 17.3 Various reflection coefficients

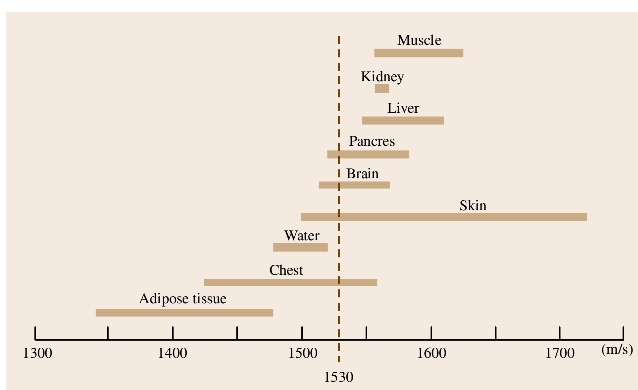

Fig. 17.4 Sound wave travel speed in tissue

Only very modern ultrasound units take into account these differences in the speed of travel of the sound waves when calculating the ultrasound images and compensate for the differences in the transit time, either manually with intervention by the user or else automatically.

17.1.5 Penetration Depth, Axial Resolution and Frequency Ranges

It is not only the number of reflecting interfaces and their reflection coefficients but also the frequency of the sound waves which influence the penetrative power of the sound waves. Higher transmission frequencies lose more energy over the same travelling distance than lower transmission frequencies and thus achieve a shorter penetration depth than lower frequencies.

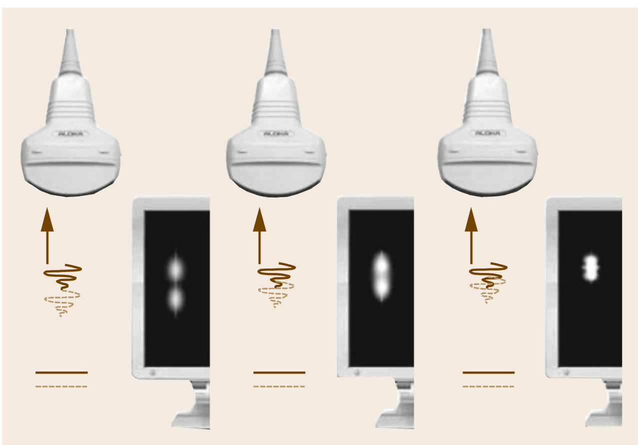

Fig. 17.5 Illustration of two reflectors at different distances

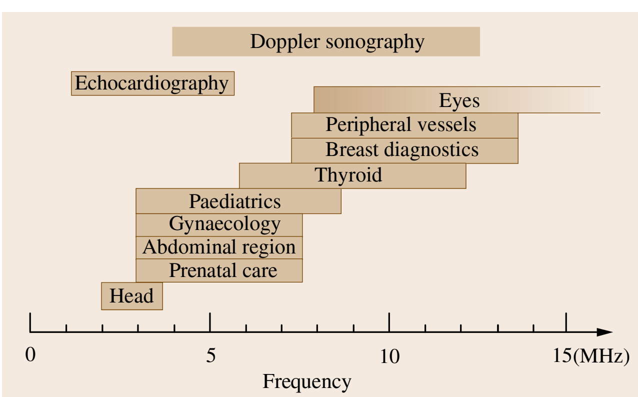

Fig. 17.6 Frequency ranges of diagnostic ultrasound

However, high frequencies have an advantage in terms of the axial resolution on account of their shorter wavelength (λ). In a theoretical best-case scenario, the minimum wavelength for separate imaging of two interfaces is a single wavelength. A frequency of 5 MHz, for example, gives a wavelength of 0.3 mm, whereas a frequency of 10 MHz corresponds to a wavelength of 0.15 mm. More individual interfaces can therefore theoretically be resolved at 10 than at 5 MHz (Fig. 17.5).

As can be seen in the figure, a pulse packet does merely not consist of a single wave with a positive and negative amplitude but also includes attack and, in particular, decay amplitudes. These lengthen the overall pulse packet by several oscillations. In modern high-end units on the market, a considerable degree of technical complexity is sought after in order to control the transient phenomena in the crystals. The aim is to generate the shortest possible pulse packets, ideally with only a single sine pulse. This can only be achieved through precise knowledge of the respective crystal characteristics of the ultrasound probe and with appropriate control using customized electrical pulse lengths and pulse forms. As a result of this *electrical behavior* of the crystals, pulse packet lengths are achieved which are nearly the size of a single wavelength. The theoretical resolution of 0.15 mm at 10 MHz can therefore approximately be achieved, as mentioned in the example above.

The axial resolution is thus proportional and the penetration depth is inversely proportional to the frequency. For practical applications of ultrasound, this provides a necessary compromise between the desire for high axial resolution and a good penetration depth.

Figure 17.6 gives an overview of the different frequencies used to produce images of different organs.

17.1.6 Influencing Factors: Pressure and Temperature

As already mentioned, the speeds at which sound waves pass through a medium are primarily dependent on the material properties of the medium. A change in the atmospheric conditions of temperature and pressure likewise influence the propagation velocity of the ultrasound pulse. Both an increase in temperature and a rise in pressure go hand in hand with faster propagation of sound waves.

In this context it is significant that the ultrasound pulse itself exerts positive and negative pressure in periodic alternation on the transmission medium. This means that the ultrasound pulse itself influences the sound-conducting properties of the medium with its pressure fluctuations. These effects on the medium and the resulting interactions continuously change the properties of the ultrasound pulse during the propagation through the tissue. This change has an effect on the signal quality and can be put to technical use.

17.1.7 Second Harmonic

During the course of time of a sinusoidal wavelength, the periodic pressure fluctuation triggered by the ultrasound pulse leads to a regularly changing sound propagation speed. The positive half-wave (pressure) moves with greater speed than the negative half-wave, with the result that the negative-going edge of the positive sound wave becomes increasingly steep as the distance from the sound source increases (Fig. 17.7).

A similar phenomenon can be observed in sea waves on the beach. The lower section of the wave, which is close to the sea bed, moves more slowly because it is braked by the ground which is getting closer all the time. The upper section of the wave is largely uninfluenced by this and thus moves more quickly. At some point, this effect becomes so great that the crest of the wave *falls* forwards, as it were, and the wave breaks.

The pulses from an ultrasound unit emitted into the body also experience this deformation on their path through the tissue. Whereas the sound wave still exhibits a virtually sinusoidal pressure curve when it is close to the crystal, as the distance from the crystal increases the shape of the wave increasingly resembles a saw-tooth. The higher the intensity of the sound emitted, the earlier and more pronounced this deformation. Every nonsinusoidal oscillation can be broken down into multiple sinusoidal oscillations of different frequencies by means of Fourier analysis (Fig. 17.8).

The saw-tooth-like shape of the changing pulse contains firstly a sinusoidal fundamental wave (a), furthermore a sinusoidal oscillation with a relatively low amplitude and half wavelength (b), and also other sine oscillations with constantly decreasing amplitudes which correspond to an integral quotient of the fundamental wavelength (c). These high-frequency components are also known as harmonic frequencies. This *harmony of sound* is familiar to us all from music, as the generation of sounds in musical instruments is subject to similar principles.

Conventional signal processing suppresses the higher-frequency signal components, so that the dominant fundamental of each reflected signal is primarily used for imaging. In the method known as *second harmonic imaging*, however, the first overtone (second harmonic) is isolated and used to obtain an image. The fundamental (first harmonic) and all other overtones are filtered out and are not used further.

Using the second harmonic frequency has a number of advantages. The main advantage is the significantly

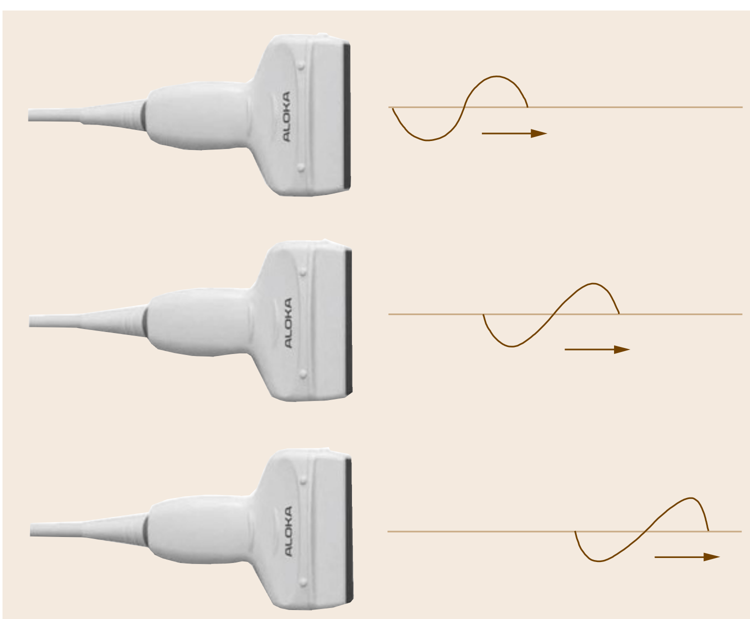

Fig. 17.7 Change in the shape of a sound wave as a result of nonlinear sound velocity

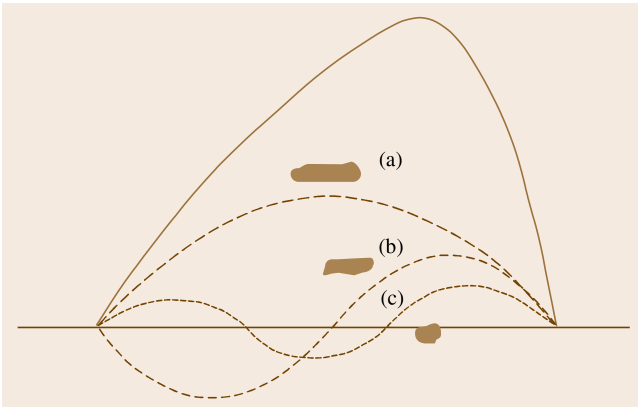

Fig. 17.8 Transformation of a nonsinusoidal oscillation into various sinusoidal oscillations

clearer differentiation between liquid and solid tissues. The reason for this is the more pronounced saw-tooth-shaped signal deformation in solid structures (tissue) and the lesser degree of deformation in liquid tissues, in which virtually no reflected harmonic waves are generated. Some companies therefore also refer to this technique as tissue harmonic imaging. Another advantage of using harmonics is that the amplitude of the interfering side lobes which occur automatically at the fundamental frequency is too low to generate overtones.

Tissue harmonic imaging is therefore a technique which reduces artifacts in the ultrasound image and makes it possible to more cleanly and clearly delimit liquid-filled cavities in particular, such as the amniotic cavity, the bladder, the cardiac cavities or cysts. These days it is also used in devices in virtually all price ranges.

17.1.8 Broadband Harmonics

One disadvantage of using the second harmonic is that the reflected second harmonic has less penetrative power as a result of the doubling in the frequency. Because these waves must cover the distance to the ultrasound probe, however, the depth of field of the ultrasound image which can be displayed is overall reduced. Use of this technology should therefore be left to the user, who differentiates depending on the organ being investigated by switching the unit on and off.

Some modern high-end devices provide another possibility for using harmonic signals, however. In addition to the harmonics already described, Fourier analysis also shows so-called half-waves (subharmonics) with half the fundamental frequency of the transmission pulse.

The simultaneous use of half-waves and second harmonics to calculate an image compensates for the disadvantage of the short depth of field because of the lower frequency of the half-wave and its better penetrative power. The reception and processing of these signals place high demands on the frequency spectrum of the ultrasound probe which can be effectively used and also on the signal processing electronics. In general, the ultrasound probe and the piezoelectric crystals used in it have optimized frequency ranges due to their material and design. The broadband capability of the ultrasound probe is indispensable for the simultaneous reception of half-waves and harmonics and is crucial to the quality of this reception. In high-end devices, this technology is found under the name broadband harmonics.

17.2 Visualization of the Blood Flow and Vascular System

17.2.1 Doppler

In addition to calculating the depth of a reflecting interface, the pulse reflection method generally used in ultrasound also provides the possibility of detecting a moving structure and measuring its speed. This is possible due to technical processing of the reflected signals, which differs from the method described before of determining the transit time and which, in addition to displaying the tissue morphology, also enables functional diagnosis of moving tissue volumes and liquids (blood, muscle, urine).

This method is named after the Austrian physicist Christian Doppler and is based on the phenomenon, which we know from everyday life, that the frequency of sound of a moving object changes according to its speed and direction. Everyone is familiar with the phenomenon of the changing sound of a passing ambulance which has its siren turned on. This phenomenon is known as frequency shifting.

The measurable change in frequency between the emitted and the reflected signal, which in ultrasound is called the Doppler shift or the Doppler frequency shift (f_D), is proportional to the speed of the moving struc-

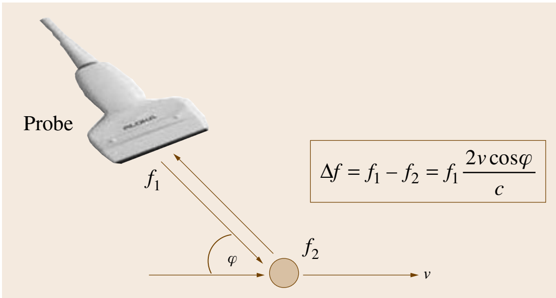

Fig. 17.9 Formula for describing the Doppler effect (f_1 transmission frequency, f_2 reception frequency, Δf measured Doppler shift, v speed of the target particle, φ angle of incidence between the sound beam direction and the direction of movement of the target particle, c sound velocity)

ture. The relationship between the Doppler shift and the speed of the target particle is illustrated in Fig. 17.9. The speed of the movement can be calculated by determining f_D in the ultrasound unit and with the aid of the known variables of sound velocity, transmission frequency and reception frequency.

A fundamental problem in the practical Doppler measurement is its reliance on angles. As can be seen in Fig. 17.9, there is a connection, dependent on the angle φ, between the moving structure and the position of the observer. Extreme cases are, firstly, direct measurement in or against the direction of movement ($\varphi = 0°$) and, secondly, measurement at right angles to the direction of movement ($\varphi = 90°$). In the first case, f_D corresponds to the actual speed, whereas in the second case the result given by the correction factor ($\varphi = 90°$, which gives cos $\varphi = 0$) does not give a measurable speed.

For practical use, it holds that an angle of between 0° and no more than 60° to the direction of movement should be used when using the Doppler technique.

17.2.2 B-Mode

In the B-mode, too, visualizations of the blood flow are also possible to a limited extent, since the reflection coefficient of blood is approximately 1000 times lower than that of tissue. On certain conditions (immediate proximity to the blood-conveying organ, slow-flow phenomenon, haematocrit, etc.), the blood flow is visualized spontaneously, which is known as spontaneous flow or spontaneous echo.

However, by subtracting two consecutively recorded image lines and subsequently superimposing them with one of the noncontrast lines, modern signal processing technology provides the possibility of imaging the movement of the reflecting structures which takes places between the two recordings. The advantage of this method (B-flow) is that it does not rely on angles. The spatial resolution of the illustration of the flow is dependent on the resolution of the B-image, however, and this can have a negative effect on the diagnosis. This method is therefore currently much less widespread than the Doppler method.

17.2.3 Ultrasound Contrast Medium

Depending on the medical indication and in order to achieve an accurate representation of the organ perfusion and the perfusion dynamics, ultrasound contrast medium is increasingly used in diagnosis. Ultrasound contrast medium is applied intravenously, passes largely unaffected through the lungs and is then pumped into the muscles and organs by the left heart in a highly diluted solution with the blood. Ultrasound contrast medium is composed of countless tiny gas bubbles with a diameter of between 2 and 4 μm and reflects ultrasound which is beamed in significantly better than the natural corpuscular constituents of the blood. There is also a second effect, which increases the visualization of the gas bubbles and therefore of the blood: the gas bubbles are considerably smaller than the spatial dimension of the sound waves.

In the compression wave of an ultrasound pulse, areas of elevated and reduced pressure periodically surround each bubble completely. In the area of elevated pressure, bubbles are compressed. Conversely, bubbles in the low-pressure area dilate. On account of these rhythmic changes in volume, the bubbles become a source of sound as they begin to oscillate. However, the acoustic pressure of the initial sound field and the change in diameter are not in a linear relationship with one another. If the pressure doubles, for example, the bubbles do not reduce in size to the same degree. As a result of this nonlinear oscillation and the slight difference in the size of the individual bubbles, they emit a broadband frequency spectrum back to the ultrasound probe. Just like the wave breaking on the beach, this broadband, nonsinusoidal signal can be understood as the sum of multiple sinusoidal single frequencies (Fig. 17.8). The second harmonic (Fig. 17.8) is usually particularly pronounced here, meaning that it can be selected in the receiving component of the ultrasound unit and used for the imaging. This method is known by the term contrast harmonic.

The so-called mechanical index (Mi) measures the mechanical effects of the sound waves on tissue and contrast medium bubbles

$$\mathrm{Mi} = \frac{p^-}{\sqrt{f}}\,.$$

Here, Mi is the mechanical index (dimensionless), p^- is the negative acoustic pressure (MPa) and f is the transmission frequency (MHz).

At a low Mi and thus a low acoustic pressure, bubbles still behave in a linear fashion, and at higher acoustic pressures their behaviour becomes nonlinear and they generate harmonics.

17.3 Equipment Technology

17.3.1 The Basic Design of an Ultrasound Unit

The fundamental components of an ultrasound unit are the devices shown in the diagram (Fig. 17.10).

The combination of these components constitutes a medical device which is approved in accordance with the respective applicable regulations, which satisfy EU and international agreements. Any change to these components or their parts (e.g. repair with a structural change, such as recoating of an ultrasound probe) means that this approval is automatically rescinded.

Ultrasound Probe

Specific ultrasound probe designs are available for different fields of use. They differ essentially in the design of the crystal array. The standard designs described in the section below can be found in different sizes, widths and radii of curvature, and in a wide variety of other ultrasound probe designs. The shape of the housing differs depending on the intended use of the respective *instrument*.

Base Unit with A/D Converter

The base unit of a modern ultrasound unit is essentially composed of two parts. The interface between

Table 17.2 Types of probes and their application

Organ	Application	Type of probe
Brain	Transcranial	Sector
Brain	Intraoperative/burr hole	Convex/sector
Pituitary	Transnasal	Linear, flexible
Thyroid, neck vessels	Transcutaneous	Linear
Heart, aorta	Transthoracic/transesophageal	Sector
Lung, mediastinum	Endobronchial ultrasonography (EBUS)	Convex
Pleura	Transcutaneous	Convex/sector
Breast	Transcutaneous	Linear
Spine	Intraoperative	Linear/convex
Liver, gall bladder, bile ducts, kidney	Transabdominal/endosonographic (EUS)	Convex
Pancreas, spleen, stomach, intestines	Transabdominal/endoscopic	Convex
Liver, spleen, pancreas	Laparoscopic	Linear/sector
Major abdominal vessels	Transabdominal	Convex
Uterus, ovaries, pregnancy	Transabdominal/transvaginal	Convex/RT3D
Bladder, prostate	Transabdominal/transrectal	Convex/linear
Scrotum	Transcutaneous	Linear
Hip	Transcutaneous	Linear
Peripheral vessels	Transcutaneous	Linear
Joints, muscles, skin	Transcutaneous	Linear

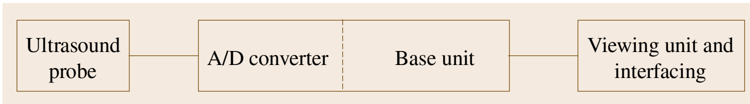

Fig. 17.10 Components of an ultrasound unit

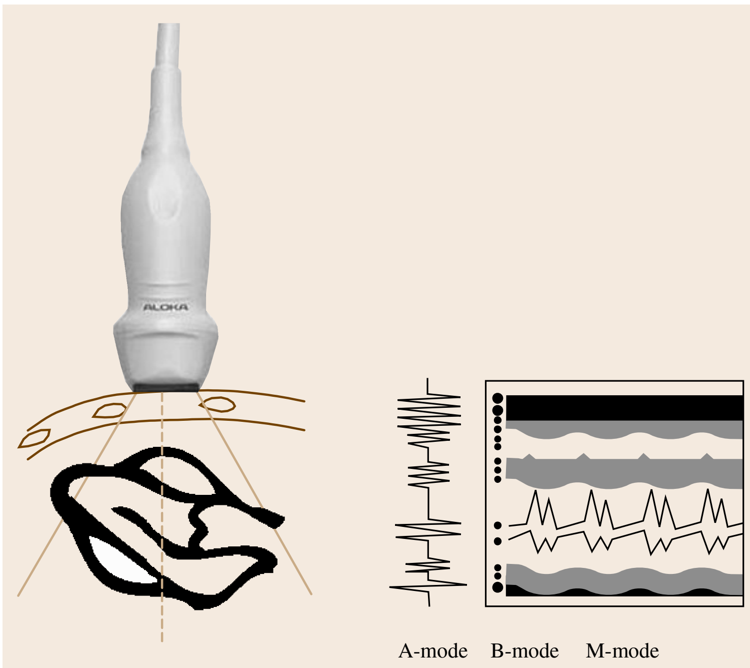

Fig. 17.11 Diagram of A-mode, B-mode, and M-mode

the analogue-functioning ultrasound probe and the digital processor is in the analogue/digital (A/D) converter. The purpose of the A/D converter is firstly to generate all of the electrical pulses and pulse sequences required for steering the sound beam and also to receive the reflected signals and convert them to digital form. It thus becomes clear that the image quality in particular is heavily dependent on the quality of the A/D converter.

The base unit also has the task of scan conversion (spatial mapping of the scan lines recorded) and further processing the digitized signals by means of image processing. Other functions include the operator-controlled computer dialogue, provision of measuring and calculation programs, as well as archive functions and generation of signals for viewing units (monitors) and interfaces.

Monitor and Interfaces

Today, digital flat-screen monitors are predominantly used. Only a few ultrasound units still work with conventional video signals. The interfaces for data transmission, printer control and archiving which are normally used today are also digital.

A-Mode

Historically, the A-mode method (amplitude mode) was the first ultrasound method used. It has been almost exclusively replaced in the medical field by the methods described below, as the diagnostic value of the A-mode is very limited. Measurements of depths and distances can only be carried out at a single point.

Along the sound propagation line, signals reflected through media boundaries are displayed as individual peaks on the depth scale depending on the transit time – the pulse reflection method (Fig. 17.11). The value of the amplitude is dependent on the ratio of the wave impedances between the media boundaries.

B-Mode

In the B-mode (brightness mode), the display of the A-mode signal is modified such that the reflected signal is no longer displayed as a peak but rather as an individual, depth-dependent (transit time) pixel. The value of the amplitude is represented by the brightness of this pixel, so that a line is generated with points of varying brightness. The simple B-mode method cannot be used diagnostically. However, it was the basis for the development of the M-mode and the 2-D B-mode.

M-Mode

The *M-mode (motion mode)* is produced by the horizontal deflection (time axis) of a B-mode line and storage and display of the resulting images. The M-mode is intended to be used for the diagnosis of moving organ parts, such as cardiac valves or cardiac muscle. It stands out in this area in particular due to its high time resolution, as just one individual line is repeatedly scanned.

2-D B-Mode

The 2-D B-mode method (two-dimensional brightness mode) is these days the most important sectional imaging technique in ultrasound diagnostics and is generally referred to as B-mode. The image is produced by quickly stringing together a number of individual B-mode lines (scanning lines) horizontally to give a flat 2-D image. Here, the image geometry is determined by the relative arrangement of the individual B-mode lines. It is also dependent on the design of the ultrasound probe. As in the traditional simple B-mode, the brightness of the individual pixels is determined by the amplitude of the reflected signals. Today, at least 256 shades of grey are required as standard.

Digitally Encoded Ultrasound

The practical applications of ultrasound require a necessary compromise between the desire for high axial resolution and a good penetration depth. Digital encoding of ultrasound pulses significantly improves this compromise, with the benefit of better penetration.

During transmission, a typical digital encoding is superimposed on the ultrasound pulse as an identification pattern. This encoding can also be detected in the ultrasound pulse after it has been reflected at the different interfaces in the patient's body, with the result that, when processing the reflected signals in the ultrasound unit, the encoded pulses can be separated from unencoded, unwanted signals (e.g. noise, artifacts). This means that only those signals which include the relevant individual code are processed further.

An example from the natural world shows that the same principle is used by bats. The individual encoding of the transmission pulses emitted by a bat allows it to identify its own pulses from a wide number of pulses from other bats. Every individual bat is therefore able, despite a large number of other pulses, to move about safely in space with its own signals.

In addition to the encoding, signal compression also takes place, making it possible to produce an encoded transmission signal with a lower level of energy, which can then form a signal with a higher level of energy again at a later point as a result of adding together the individual digital pulses.

Two fundamental advantages of digital encoding emerge from this:

1. Unwanted signals are suppressed.
2. The useful signal is amplified.

Unwanted signals such as noise and artifacts are suppressed, and at the same time the useful signal is amplified and raised above the ambient noise level. At the same frequency and thus also at the same resolution, the effects mentioned bring about an increase in the penetration depth. This significantly reduces the problem of achieving a high resolution whilst at the same time obtaining a high penetration depth. This method does not put us in a position to be able to do away with the compromise between frequency (resolution) and penetration depth. A significant improvement in the situation is possible, however.

17.3.2 Doppler Ultrasonography

As outlined in the introduction to the fundamental principles, Doppler ultrasonography is used to detect blood flow. Different methods can be used technically:

- Pulsed-wave (PW) Doppler
- High-PRF Doppler
- Continuous-wave (CW) Doppler
- Color Doppler.

With Doppler ultrasonography, a distinction is also drawn on the basis of whether a device operates solely in the Doppler mode or whether it operates in the so-called duplex mode by superimposing a B-mode. In *duplex mode*, the scanning point (*sample volume*) at which the velocity of blood flow is measured is displayed in B-mode. The anatomical mapping at the point of the velocity measurement is therefore defined precisely. A Doppler device for use in duplex mode must also include all the technology for generating a B-mode.

Fig. 17.12 Duplex Doppler, B-mode with sample volume (*left*), and PW spectrum (*right*)

PW Doppler

The pulsed-wave (PW) Doppler technique uses the pulse reflection method and also the A-, M- and B-mode imaging methods. The processing of the transit time information is used here to determine the sample volume position. Reflected signals from the sample volume are not used to display an image but rather to measure the velocity according to the Doppler principle.

Similar to the M-mode, the time profile of the velocity of blood flow is displayed on the monitor. The position of the sample volume can be changed and positioned accurately by sight in duplex mode (Fig. 17.12).

The maximum velocity of blood flow which can be measured using PW is limited by the repetition frequency of the pulses (PRF). By using other, additional sample volumes, this limitation – which results from what is known as aliasing – is shifted into a velocity range which is no longer diagnostically relevant.

High-PRF Doppler

The PW Doppler technique is limited by what is known as *aliasing*. Above a certain frequency and velocity of blood flow, this causes incorrect display (alias display) of the Doppler signal, which can lead to incorrect interpretation of the flow direction.

This effect is dependent on the frequency of the pulse (scanning frequency or pulse repetition frequency (PRF)), which should be as high as possible. An artifact such as this can be observed in Western films, for example, when the carriage wheels suddenly appear to rotate backwards because the scanning frequency (television frame frequency) is too low.

The high-PRF Doppler technique which should be found in all modern duplex systems was developed to avoid this effect of aliasing while maintaining the depth selectivity. It allows the aliasing cut-off frequency to be shifted upwards by increasing the PRF. The price paid for this is multiple sample volumes, which results in a slight restriction in the depth selectivity.

The use of duplex systems with automatic switching from PW to high-PRF is a useful addition to the pure PW mode.

Today, duplex devices should include frequency analysis in order to make it possible to detect pathological flows in the characteristics of the Doppler spectrum displayed. The Doppler signal is broken down and displayed in its individual frequency components by Fourier transform or similar methods (Fig. 17.8).

When the angle of incidence is known, frequency components can be equated with velocity components. Further information about the velocity is thus obtained. The most important here is the information about the degree of turbulence, which gives insights into pathological flow.

CW Doppler

The CW Doppler (continuous wave) technique is the only ultrasound method which is not a pulsed method. Two physically separate crystals, a transmission crystal and a reception crystal, are accommodated in a probe and operate simultaneously. Whereas one crystal constantly transmits, the second crystal continuously receives the reflected signals. Alternatively, it is also possible to use groups of crystals, e.g. in a phased array ultrasound probe.

Due to the continuous transmission, depth mapping is no longer possible, however, as is it is not possible to measure pulse transit times. The velocities measured are rather divided along the entire path of the measuring beam. The resulting advantage is the ease with which it is possible to measure high velocities of blood flow, such as those which are to be found in high-grade stenoses. The disadvantage of this method compared with the PW Doppler technique is the lack of depth selectivity.

Color Doppler

The industrial development of the first color Doppler in Japan (ALOKA, 1985, SSD-880) was an important step in the development of ultrasound diagnostics.

The color Doppler is in principle a PW Doppler. Here, it is not just a single sample volume which is used but several hundred. These sample volumes are combined in lines to form an extensive two-dimensional Doppler image. The Doppler image is superimposed in the correct position on the black-and-white B-mode image. These two images can be viewed together in real time in the real-time mode.

In order to be able to distinguish the Doppler information from the morphological black-and-white information (B-mode), it is displayed with color-coding (Fig. 17.13).

A generally accepted illustration of the direction of flow is given with the colors red (towards the ultrasound probe) and blue (away from the ultrasound probe). However, it must be mentioned that there is no additional standardized color coding system. In the individual color Doppler systems, too, various color coding systems (color charts) can be accessed which more or less follow subjective preferences.

The effect of aliasing mentioned above occurs as an artifact in the color Doppler technique, too. This has the effect that, in the middle of a flow which is color-coded blue, for example, a red spot appears which is completely surrounded by blue (Fig. 17.14). This can be seen when the maximum velocity which can be displayed is exceeded at precisely this point. The color then *turns* into the other color.

As it is obviously impossible for a *red island of reverse flow* to suddenly appear in the middle of a blue flow, the user can easily identify this phenomenon as aliasing.

The velocity (cut-off frequency) above which aliasing occurs is called the Nyquist limit

$$\text{Nyquist limit} = f_{\text{Ny}} = \frac{\text{PRF}}{2} .$$

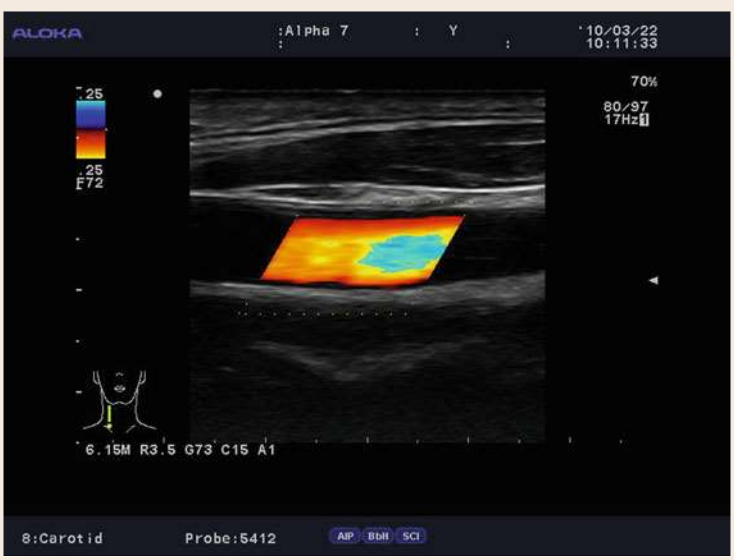

Fig. 17.13 Color Doppler with aliasing

Fig. 17.14 Vascularization in the fetal brain, imaged using eFlow technology

Aliasing can even be useful here in that it shows us the point in the blood flow with the highest velocity. In order to measure the highest velocity, we position the sample volume at this alias position. Using the color Doppler technique it is then possible to detect and pinpoint the extent of pathological flows such as jets and to locate small vessels in real-time mode.

If the color Doppler technique is also added to the duplex mode so that the B-mode, color Doppler and Doppler spectrum are all displayed, this is referred to as the triplex mode.

Power Doppler

The term *power Doppler* is understood as meaning another version of the color Doppler technique. It is also known by the following synonyms:

- Amplitude Doppler
- Color angio
- Power color, etc.

The power Doppler technique also uses the Doppler shift to image blood flow. However, it does not display the velocity of the flow using color-coding, but merely shows the amplitude of a point of movement in the received signal detected by the Doppler shift present. The amplitude represents the number of reflectors (mainly erythrocytes) and is not displayed in the normal color Doppler technique. The power Doppler is generally more sensitive and can more easily detect slow-moving blood particles. It therefore fills the vascular lumen better than the normal color Doppler is able to, as the low amplitudes of the available signal information are not sufficient for further analysis with the normal color Doppler.

High-Resolution Modern Color Doppler Techniques

At this point it is worth mentioning a further development of the power Doppler technique which is known by various different names (e.g. eFLOW, dynamic flow, etc.). In some modern units, the method can be found with markedly higher spatial resolution values (<0.3 mm). It is extremely well suited for accurate, two-dimensional illustration of the vascular lumina through which blood is actually flowing. This method is of advantage particularly in the case of slow velocities or small vessels and when imaging a complex organ vascularization, but also in the case of stenoses.

It is made possible using modern pulse generators (beamformers) which shorten and optimize transient phenomena in transmission pulses. This makes extremely short pulses possible, which enable a higher spatial resolution.

Tissue Doppler Imaging

A special Doppler technique has become established for imaging the heart, and this technique does not color-code the blood flow but rather the cardiac tissue, which is to say the cardiac wall (myocardium). This is referred to as *tissue Doppler imaging* (TDI). Since the move-

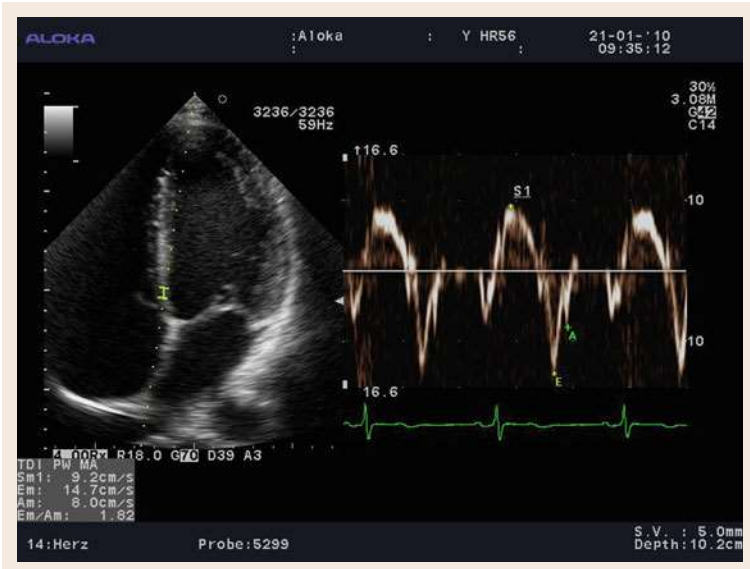

Fig. 17.15 Tissue Doppler imaging (TDI)

ment in the myocardium is considerably slower than that of the blood, low-pass filters are necessary here instead of the high-pass filters which are customary in the blood flow Doppler. This technology enables conclusions to be drawn regarding the velocity of individual segments of myocardium (Fig. 17.15) and thus allows specific diagnosis in the case of disorders concerning the movement of the wall, e.g. as a result of a myocardial infarction or diastolic dysfunction. Additional information can be found in the further reading.

17.3.3 Types of Probes

The types of probes used today are primarily divided into two groups. Mechanical probes and electronic probes, or combinations of the two, are almost exclusively used.

Mechanical Probes

Today, these are still used for special applications (e.g. radial technique, skin diagnostics). Here, the deflection

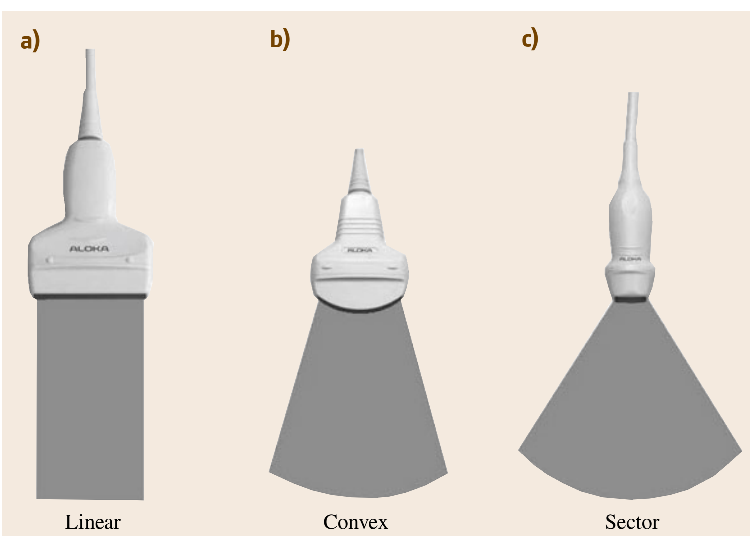

Fig. 17.16a–c Sound field geometry of various types of probes

for generating a two-dimensional image is carried out by a small electric motor in the ultrasound probe.

Electronic Probes

Electronic probes are used far more frequently, and we know these as linear, convex and sector scanners. They differ in principle due to their deflection geometry and their deflection method (Fig. 17.16).

The deflection of the ultrasonic beam is realized electronically by time-shifted activation of a large number of crystals; no mechanical wearing parts are used whatsoever. The maximum transmission frequency is usually approximately 15 MHz.

Linear Probes. The simplest form of an electronic probe is the linear scanner. It consists of a large number of crystals arranged next to one another, and depending on the size of the probe it can contain up to approximately 400 crystals (Fig. 17.17).

A scanning line is always constructed using several crystals, and in this case six crystals lying next to one another are used, for example. This group of active crystals is also called an array, which has given us the frequently used term linear array. Each crystal is activated by an electronic pulse and transmits a sound wave which merges with the others to form a wave front. As a result of various physical effects, a sound field with a focal constriction is generated in the typical sound field geometry illustrated (solid line). Another six crystals are then activated, forming the next (dashed) sound field. According to the same principle, the next (dotted) sound field is then generated in turn, and this is repeated in the direction of the arrow until a complete image can thus be constructed from the reflected pulses. In order to generate a real-time display with 20 images per second (20 Hz), all of the processes described must have taken place within approximately 1/20 s.

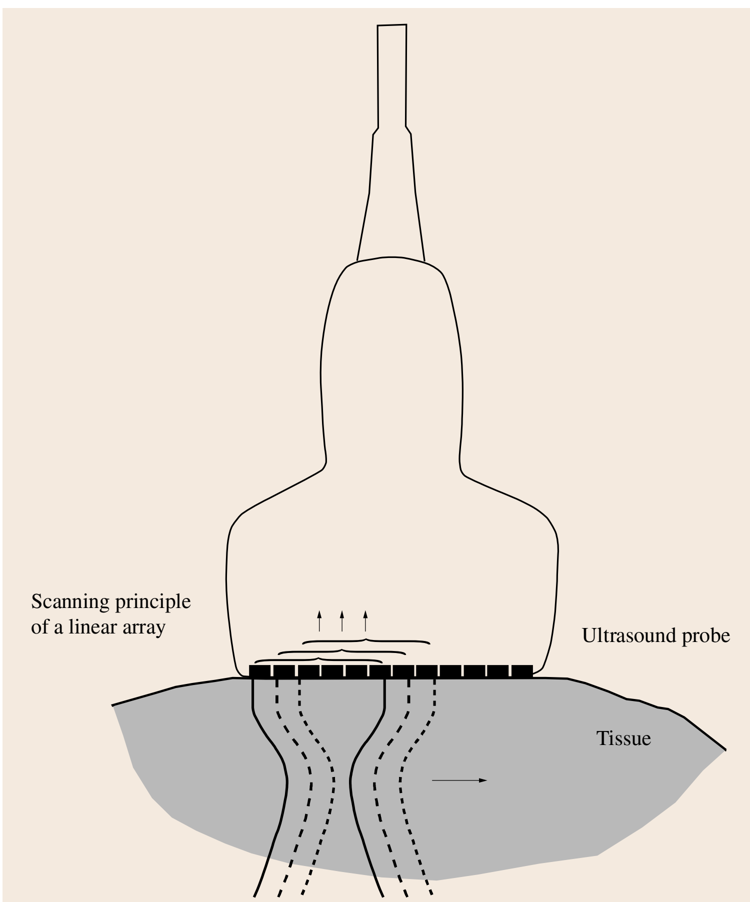

Fig. 17.17 Scanning principle of a linear probe

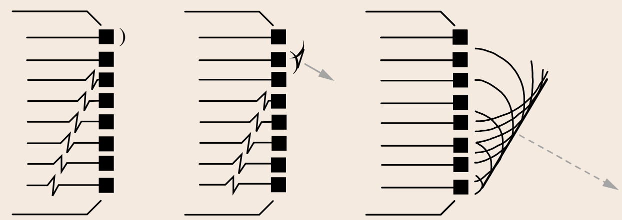

Fig. 17.18 Phased array principle

Convex Probes. In the *convex probe*, also known as a curved array, the same method is used. The only difference here is that the crystals are not arranged on a linear contact surface but on a convex contact surface. Different radii of curvature are available for different fields of use (Fig. 17.16b).

Sector Probes. In the electronic sector, also called a phased array, considerably fewer crystals are used than in the linear or convex array. The electronic activation is carried out with a time shift or phase shift, which is where the term phased array comes from. However, this is no longer done from the outside in but continuously from one side to the other, with the result that an oblique wave front is generated (Fig. 17.18). Changing the activation at an appropriate speed achieves an oscillation in the beam, which in turn produces a two-dimensional image in the form of a sector.

The advantage of sector probes can be clearly seen from the example of avoiding rib shadows (Fig. 17.19), as the small contact surface of the sector probe enables examination through the intercostal spaces.

A disadvantage of the sector technique is caused by the divergence of the ultrasonic beams at greater depths, which results in the lateral resolution becoming increasingly poor in deeper tissues. The gaps which appear are filled in by the device using interpolations.

17.3.4 Focusing

The focal constriction, mentioned above, is important in all electronic probes, and this is known as the focal area. The narrower the sound field in this focal area, the better the lateral resolution of the system, that is to say in the lateral direction, perpendicular to the sound propagation direction. The diameter of the ultrasonic beam in the focal area depends on the different electronic focusing techniques of the device. Resolution values cited by the manufacturers usually relate to the focal area. The focal position is also crucial, however. Ideally, the focal position should be precisely where the organ to be examined is located. This is where the biggest advantage of electronic probes over mechanical probes is to be found: the possibility of shifting the focal position electronically.

Dynamic Focusing

In dynamic focusing, the crystals are no longer all activated simultaneously; instead, the outer crystals are activated first, followed by the central crystals. The position of the focus is determined by the time-shifted activation between the outer and inner crystals (delay line). In addition, in electronic probes it is possible to use multiple focal zones simultaneously, which makes it possible to produce a sharp image over virtually the entire depth of field, although this comes at the expense of the image frequency. This is called dynamic focusing (Fig. 17.20). The image regions of the individual focal zones are recorded one after the other using a buffer memory and are finally combined to form a complete image which is focused throughout (Fig. 17.20b).

Modern devices have up to ten focal zones, but only four of these are usually used simultaneously for dynamic focusing because, as already mentioned, the use of a buffer memory reduces the image frequency. When using conventional array technology, this focusing technique is only possible in the longitudinal direction of the ultrasound probe, however, and only affects the lateral resolution.

Slice-Thickness Focusing

At right angles to the ultrasound probe, and thus also at right angles to the section plane, the scanning lines are only focused at a depth which is dependent on the shape of the crystals. Both at close range and at distant range, the ultrasound can propagate relatively uncontrollably. The actual two-dimensional sound field therefore also extends in an unwanted fashion at right angles to the section plane. This unwanted extent is called the slice thickness.

As a result, very small vessels, cystic hollow bodies or low-echo lesions are not intersected exactly in the middle, and every scanning beam also records echogenic structures which are behind or in front of the small structures. Other echo signals from exactly this depth therefore impinge on the ultrasound probe, and these cannot unambiguously be attributed to the small structure (Fig. 17.21a). Only in the focal area is the extent of the slice thickness of the sound field so small that no tissue parts outside of the small structure are affected.

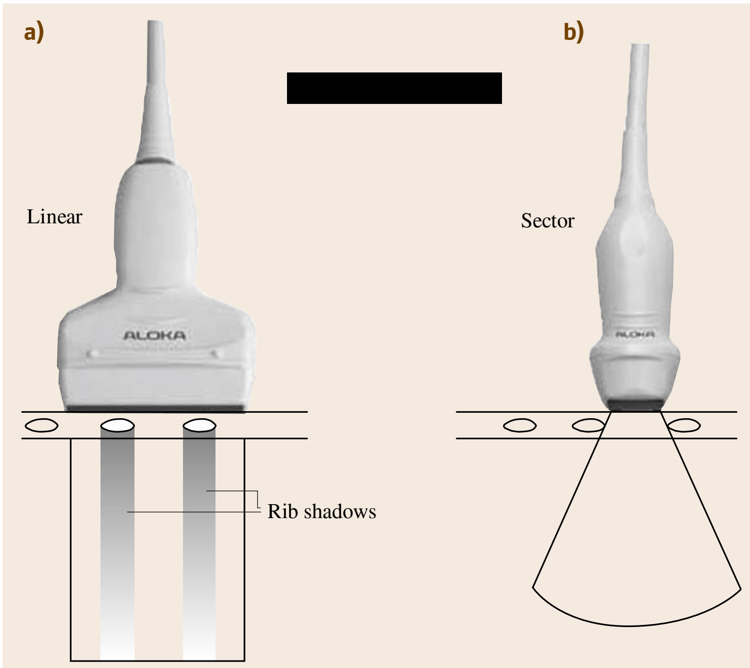

Fig. 17.19a,b Comparison of sector and linear probes

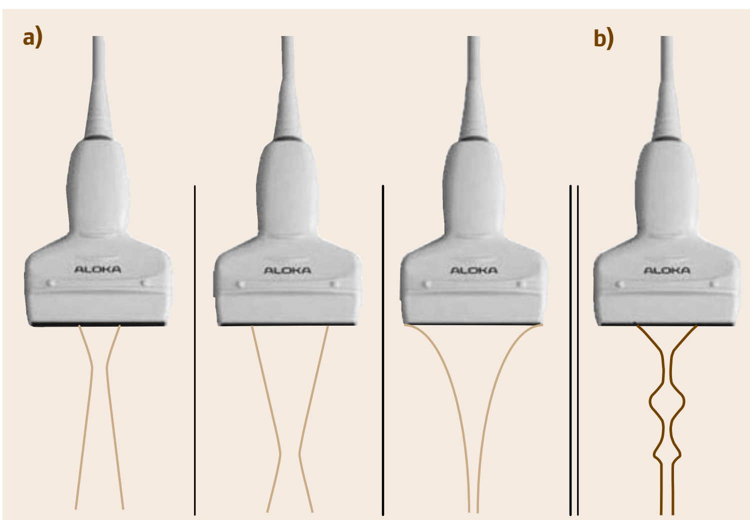

Fig. 17.20a,b Principle of dynamic focusing

The matrix array technique is a new ultrasound probe technique which also provides electronic focusing at right angles to the section plane and thus provides a reduced and uniform slice thickness from close range to about 25 cm.

Conventional ultrasound probes consist of a single row of piezoelectric crystals lying next to one another and therefore also only permit focusing in this single lateral plane. The matrix array ultrasound probe has a two-dimensional arrangement of up to approximately 1000 tiny square crystals which can be activated selectively. Electronic focusing as described by time-shifted activation of the individual elements is therefore not only possible in the lateral plane but also perpendicular to it (elevation plane). This technique largely eliminates slice thickness artifacts (Fig. 17.21b) and also enables

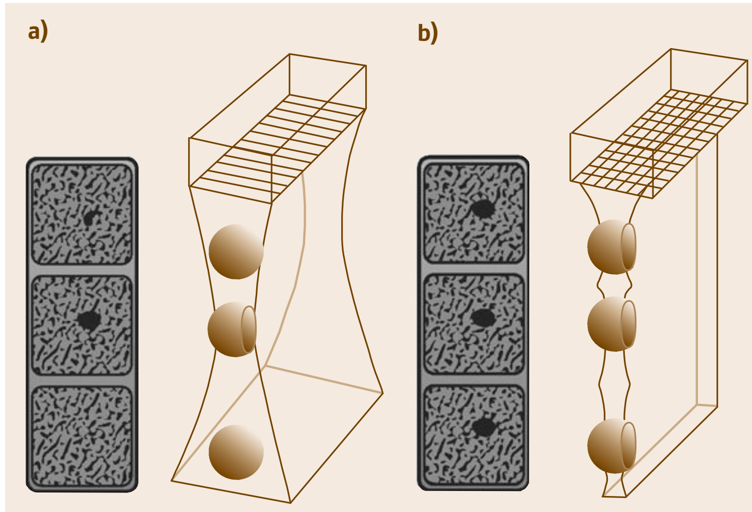

Fig. 17.21 (**a**) Conventional technique versus (**b**) matrix array

very small structures to be visualized without artifacts and with a high resolution.

Another positive effect of the matrix array technique is the considerable improvement in the penetrative power on account of the reduction in beam divergence and thus the increase in the energy density which accompanies it at the same time. It allows higher ultrasound frequencies to be used while maintaining the same penetration depth, thus enabling an improvement in the frequency-dependent axial resolution.

17.4 Three-Dimensional Ultrasound (3-D, Real-Time 3-D)

The development of three-dimensional ultrasound technology took into account the needs of many users for three-dimensional imaging of anatomical structures. The objective was among other things to simplify and improve the diagnosis of structures which can often only be assessed with difficulty in 2-D images.

17.4.1 Acquisition Techniques

Modern ultrasound diagnostics can also record and display three-dimensional blocks of data. To this end, a third dimension perpendicular to the 2-D plane is added to the two-dimensional (B-mode) ultrasound images and processed as a 3-D data set.

In everyday clinical practice, 3-D technology was only able to establish itself with the development of efficient computer processors, which enabled a true real-time 3-D display. When 3-D technology emerged in the 1980s, computing times were still in the region of hours. As a result of precisely this real-time capability of the systems, 3-D technology has recently also become known and marketed as 4-D ultrasound.

Hands-Free Technology Without Position Sensors

This simple technology uses the conventional 2-D ultrasound probe, which is moved or panned as uniformly as possible by the examiner parallel to the 2-D plane or around a fixed point in order to record an image sequence. The number of images to be recorded which will later be used to reconstruct the 3-D data set is specified before the procedure is begun (Fig. 17.22). This hands-free technology does not allow accurate three-dimensional mapping, however, as the distance actually recorded is not known and there is no guarantee of a precise and straight recording movement. This also means that no measurements can be made and volumes cannot be determined.

Hands-Free Technology with Position Sensors

This hands-free technology is a development of the technology without position sensors and is used to avoid the disadvantage of undefined recording. By attaching a position sensor to the 2-D ultrasound probe, constant

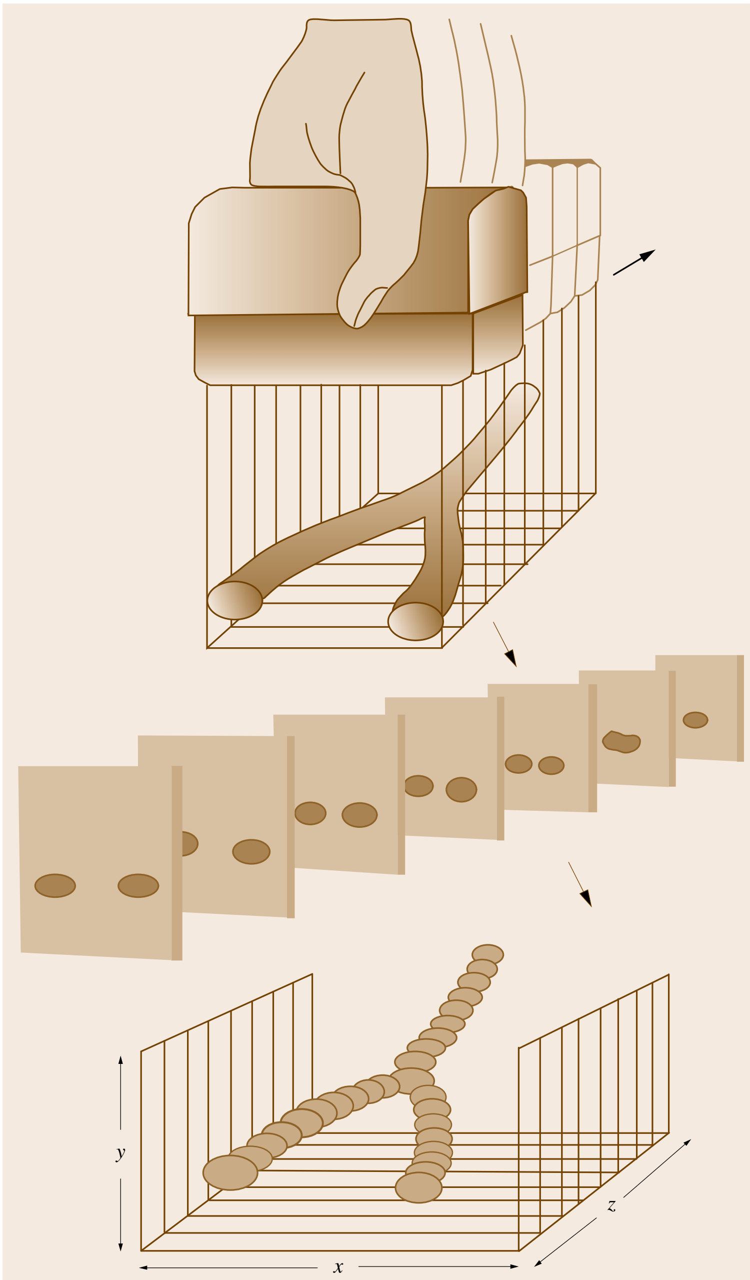

Fig. 17.22 Data block from multiple parallel sectional images using 3-D hands-free technology

determination of the position of the ultrasound probe and thus spatial mapping of the individual 2-D images becomes possible. Data sets recorded in this way allow measurements to be made and volumes to be determined.

The position detection can be achieved using different techniques and position sensors. In addition to sensors in the electromagnetic field, infrared, ultrasound and acceleration sensors (gyro sensors) are also used. These forms of sensor technology all have their own

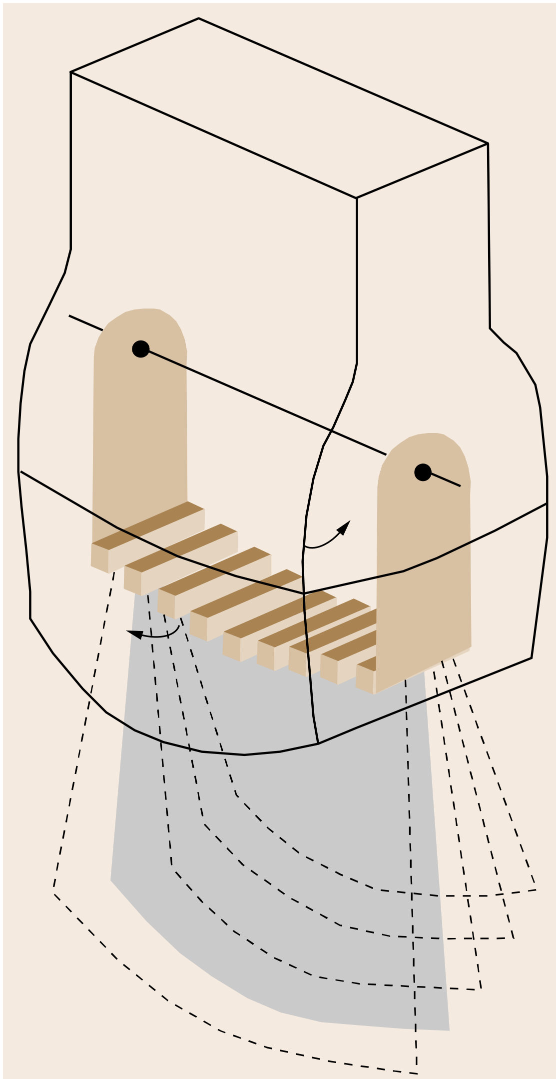

Fig. 17.23 Integrated 3-D/4-D ultrasound probe with mechanical deflection

methodological advantages and disadvantages. The disadvantages predominantly have an adverse effect on the spatial resolution.

Integrated 3-D Ultrasound Probe with Motorized Deflection

In order to overcome the problems described with hands-free technology and ensure simple handling, motor-driven 2-D crystal arrays have been combined with mechanical displacement transducers which enable reproducible recording of a 3-D data block. The array oscillates in a special liquid-filled ultrasound probe housing and scans the anatomical structures repeatedly at such a speed that a real-time representation of the volume can be calculated (Fig. 17.23).

This type of ultrasound probe is currently primarily used in gynaecology to image the foetus, as it achieves very good resolution with integrated convex arrays.

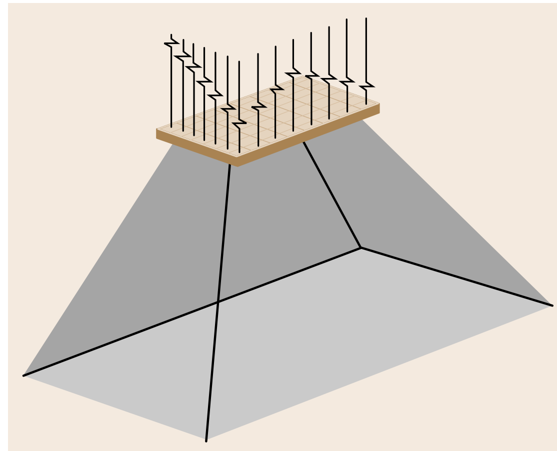

Fig. 17.24 3-D frustum of a pyramid, generated with a matrix phased array ultrasound probe

Integrated 3-D Ultrasound Probes with Electronic Deflection

This smaller and lighter construction manages without mechanically moving components. As with the matrix array ultrasound probe, the array of these probes consists of a two-dimensional, usually square arrangement of tiny piezoelectric crystals which can be activated selectively. According to the principle of the sector phased array method, the crystals are electrically activated at different points in time, so that the scanning beam forms a narrower or wider angle with respect to the array surface. By time-shifted activation of the respective crystal rows and columns, unlike with the conventional phased array ultrasound probe the direction of the beam can be determined in both horizontal axes (Fig. 17.24).

This type of ultrasound probe is currently used primarily in cardiology, as it is predestined for intercostal use due to its small contact surface.

17.4.2 3-D Reconstruction

Irrespective of the scanning method, in all 3-D methods all of the sectional images acquired are lined up during

Fig. 17.25 Surface 3-D

the scanning process such that their positions and angles are correct, providing a three-dimensional data set.

This spatial data block represents the basis for all further calculations. Different display calculations are used here:

- Surface display
- Multiplanar display
- Transparent display.

Surface Display

Surface display requires a considerable difference between the wave impedances of the tissue structures (change in impedance), which generally exists between liquid and solid structures. In order to visualize the desired interfaces, it must be possible to clearly delineate the object under examination from its surroundings.

These conditions can often be found when performing diagnosis during pregnancy, as the skin of the foetus is demarcated from the amniotic fluid by a considerable change in impedance.

Specific algorithms for representing surfaces in 3-D data sets (rendering) and further calculations from the graphical image processing, such as the generation of artificial shadows, for example, allow a display which is photorealistic (Fig. 17.24).

Multiplanar Display

This technology allows the examiner to image section planes which cannot usually be displayed from an application point of view. It is possible to simultaneously display multiple sections through an organ or structures which are at any desired angles with respect to one another. The examiner thus gains the possibility of better visualizing anatomical or pathological structures spatially, measuring them and determining their volume.

Transparent Display

By using filtering, it is possible to suppress faintly reflecting structures to the advantage of structures which reflect signals strongly. This allows clear 3-D imaging of bones, for example, which is usually necessary when searching for anomalies in the foetal skeleton.

17.4.3 New and Additional Technologies

This section gives an overview of particular new technologies which have started being used in recent years in ultrasound technology:

- Extended field of view
- Trapezoidal technique
- Steered compound imaging (real-time compound imaging)
- Speckle reduction
- eTracking.

Extended Field of View

The term extended field of view (EFOV) is understood as meaning the extension or widening of the 2-D

Fig. 17.26 Scrotum imaged using extended field of view

imaging plane. By shifting the ultrasound probe in its longitudinal axis, an extended two-dimensional image is generated.

This does not involve extending the information in the 3rd plane (perpendicular to the 2-D plane) as is the case with 3-D, but rather extended the 2-D plane itself. Using image processing algorithms, matching sections of the respective individual images are lined up together using pattern recognition, and these individual images therefore produce a calculated, new and coherent picture of the organ structures (Fig. 17.26).

This technology is also known by other trade names: SieScape, Freestyle, LOGIQ View, Panoramic Ultrasound, etc.

Trapezoidal

With the trapezoidal technique, the width of the sound field of a linear probe is extended. By time-shifting the activation of the outer groups of crystals (phasing or steered beam principle), a trapezoidal sound field is generated (Fig. 17.27).

This technology is known by other names, such as Virtual Convex, for example.

Steered Compound Imaging (SCI)

The conventional electrical activation of linear and convex array ultrasound probes has the effect of emitting a transmission pulse at right angles to the surface of the ultrasound probe. Anatomical structures or interfaces running parallel to the ultrasonic beam are struck by the ultrasonic beam either tangentially or at a very acute angle. Under these conditions, the portions of the ultra-

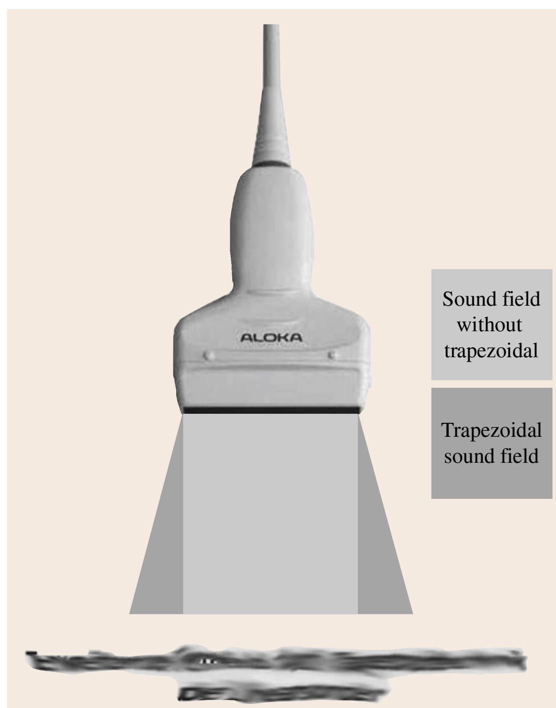

Fig. 17.27 Extension of sound field using trapezoidal technique with linear probes

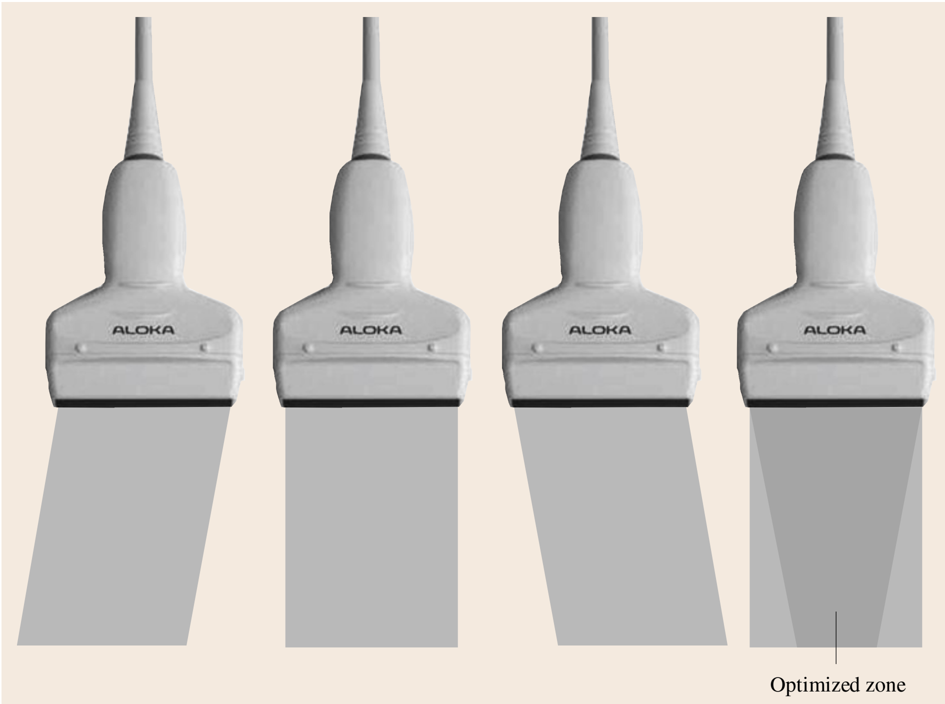

Fig. 17.28 Image optimization using steered compound imaging

sonic beam reflected in the direction of the ultrasound probe are very low. These structures therefore appear only faintly or not at all. The scanning beam also barely reaches the structures lying distal to these interfaces, because the majority of the energy of the ultrasonic beam is deflected by reflection.

In addition to the known lateral edge shadows artificial inhomogeneities also appear in the distal structures and interfaces. The consequences of these phenomena can be reduced if the structures being examined are sounded not just from one direction but from different directions using beam steering (phasing) and the resulting sound reflections are superimposed using image processing to form a compound ultrasound image. Usually, three consecutive images are generated, which have been recorded in a range between −30, 0 and +30° (Fig. 17.28). Modern devices furthermore also allow angle steering, adapted according to requirements, in various intermediate stages which makes it possible to image different depths of field. Every reflecting interface is therefore optimally sounded from different directions.

This technology is also known by other trade names, such as Realtime-Compound-Imaging, SonoCT, Cross-Beam, etc.

Speckle Reduction

Because of their physical properties, ultrasound images are always accompanied by noise. The signal-to-noise ratio is in the region of approximately 1.9 and can only be visually and subjectively improved by corresponding image processing. As a result of the constantly increasing computational power of computers, further image-processing steps can be added after the actual acquisition of signal data. Special algorithms allow image characteristics to be emphasized and changed. This includes image homogenization and the accompanying subjective noise suppression, and also, for example, edge enhancement of adjacent organ structures.

These image-processing measures can in many cases aid fast and easy diagnosis, but they are not always and exclusively of use, as both false positive and also false negative image information can be generated. The trained operator should therefore always be free to decide on their use and also their degree of use.

Vessel Distension Analysis (Echo Tracking, eTracking)

The growing awareness of relationships between cardiovascular diseases and vascular function and also

Fig. 17.29a,b Vessel distension analysis using eTracking

vascular mechanics has meant that diagnostic parameters can be raised by means of a series of different technical methods (ultrasound, tonometry, plethysmography, etc.).

Ultrasound can be used to calculate a number of relevant parameters, such as the pulse wave velocity (PWV), vascular elasticity (β-Index), augmentation index (AIx) and arterial function (FMD), from the vascular distension. The distension is determined in real time by analysing the raw data signal. Using specialist software, the movement of moving organ structures is followed by user-defined points along an ultrasonic beam (tracking). This is done with a high level of temporal (1 kHz) and spatial precision (1/16 of the wavelength).

A development of this analysis method is based on the simultaneous and colocated detection of the velocity of the blood flow using Doppler technology. This allows further parameters to be calculated, e.g. pulse wave propagation or wave intensity, which describe the coupling between the heart and the vascular system downstream (Fig. 17.29a,b).

17.5 Operation of an Ultrasound Unit

17.5.1 General Conditions

Operation of an ultrasound unit has today become standard in most medical fields. Some important general conditions should be observed and are detailed below.

Documentation

The operator of an ultrasound unit is generally required by the accounting body and by legal authorities to document findings.

All suppliers of ultrasound equipment provide relevant advice and solutions on request for the analogue and digital storage of data and images for the findings. In the simplest scenario this includes thermal printers or internal digital memory with CD/DVD/USB storage or export interfaces, such as the vendor-neutral DICOM (digital communication in medicine) standard for network-based archiving of still images or image sequences and measurement data.

The DICOM standard not only governs the archiving of medical image data as well as other modalities such as CT and MRI, but also provides further functions and procedures which support work-flow (e.g. work list, print, structured report, etc.).

Application-Specific Additions and Upgrades

Modern digital ultrasound units allow additional hardware or software to be installed to extend the scope of use of the unit. This includes, for example, units for simultaneous ECG recording, memory upgrades, specific analysis programs (stress echo, contrast medium assessment, TDI evaluation, analysis of the vessel distension and pulse wave velocity, etc.).

In addition to these device-specific upgrades, there is also always the possibility of extending the intended use of a device with additional ultrasound probes.

Maintenance and Repair, Updates

The ultrasound unit manufacturers provide recommendations for safe operation and warnings against improper use of units and probes. The user should find this information in the safety instructions provided with the equipment documents.

Important: in line with the Medical Devices Operator Ordinance (Medizinische Betreiberverordnung, MedBetreibV), the operator of an ultrasound unit has certain duties to ensure the safety of the systems used. To this end, the ultrasound unit manufacturers and retailers offer tailored maintenance and inspection agreements which conform to the law support service for their customers. These contracts often also include the option of insuring against damage to the units and probes. There are also offers for regular updates of the device software.

Hygiene

In general, the operator should inform him or herself about possible legal hygiene regulations in the context of the planned examinations. In addition, the guidelines of the professional associations include recommendations regarding hygiene which should also be observed.

For the cleaning of units and also the cleaning, disinfection and sterilization of ultrasound probes and auxiliary equipment, such as puncture adapters, the manufacturers provide information about the methods to be used and approved cleaning, disinfection and sterilization agents.

Safe Use of Ultrasound Diagnostics

According to current scientific knowledge, the diagnostic use of ultrasound which complies with standards does not have any unwanted side effects. However, it is recommended that the Doppler mode be used cautiously, particularly for the purpose of foetal examinations. Because ultrasound energy is absorbed and converted into heat at the transition between tissues in bony structures, it is not possible to rule out cell-damaging effects with certainty. It is therefore imperative that the recommendations of the relevant professional associations are observed.

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