Chapter 11
The Physics of Ultrasound and Some Recent Techniques Used

Gert Karlsson

This chapter will focus on understanding the underlying physics of diagnostic ultrasound. This will help ultrasound operators get the best possible performance and understand how simple adjustments can improve the quality of the scanning thus obtaining a better diagnostic tool. To achieve a good performance of the scanning, the examiner needs to know how the adjustments can be used to obtain the best possible diagnostic image.

It is also the aim to go through some frequently seen image artifacts and how to reduce the negative influence of these artifacts, thereby ensuring that the ultrasound can be a real help in the diagnostic session. Finally some more recent technical developments that can influence the future choice of equipment are outlined.

The Piezoelectric Effect

The operation of all ultrasound transducers is based on the piezoelectric effect. The materials used for making crystals for transducers are endowed with a property called the piezoelectric effect.

This effect means that when the material is submitted to a voltage, the piezoelectric material deforms slightly. Inversely, if the material is deformed by external forces, a small voltage change results.

In diagnostic ultrasound, short-duration electrical pulses are applied to the ultrasound crystals during the transmit phase giving rise to short-duration deformations of the crystals. These deformations result in an ultrasound transmit waveform being transmitted into the tissue with a rather constant velocity, estimated on average to be 1,540 m s^{-1} in human tissue. (The velocity is much lower in air and much higher in bone.)

Characteristics of Ultrasound

Ultrasound is characterized by its frequency. The frequencies of ultrasound used for diagnostic purposes in urology are usually in the range from 2 to 20 MHz.

A frequency of 2 MHz means that the ultrasound wave generates 2,000,000 cycles per second.

The lower the frequency of ultrasound, the greater its ability to penetrate deeper into tissue, but because the wavelength becomes longer with decreasing frequency, the resolution will be lower.

The higher the frequency of ultrasound, the poorer its ability to penetrate into deeper parts of the tissue, but because the wavelength becomes shorter with increasing frequency, the resolution will be higher.

Due to this relationship between resolution and penetration, the general rule for ultrasound scanning is that the frequency used should always be as high as possible, taking into account how deep in the tissue the target organ is situated – in other words, how much penetration depth is desired. For very superficial organs, like testis and penis, very high frequencies (>10 MHz) are used. For organs like the kidneys, more penetration is needed; therefore, lower frequencies (3.5–5 MHz) must normally be used.

For prostate scanning, the change in procedure in the last decades (from scanning from the outside, through the bladder, to scanning the prostate using transrectal ultrasound (TRUS)) has meant that much higher frequencies are now used (6–12 MHz instead of 3–5 MHz).

The result of this is a significantly higher image resolution during ultrasound scanning of the prostate.

The Principle of Ultrasound Scanning

The ultrasound waves used in urology are transmitted as a series of short pulses with a duration of a few microseconds. Between these short pulses being sent out, the transducer is receiving the echoes coming back from the different depths of the tissue. The time necessary for receiving an echo depends on the tissue depth.

Ultrasound is reflected when it goes from one kind of tissue to another. How much is reflected depends on the change in impedance between the two kinds of tissue and on the angle

O. Ukimura and I.S. Gill (eds.), *Contemporary Interventional Ultrasonography in Urology,*
DOI: 10.1007/978-1-84800-217-3_11,

of incidence of the ultrasound beam. The reflected part of the ultrasound energy is seen as echoes of different brightness on the ultrasound image.

If most of the incident ultrasound wave is reflected by some structure, the echo on the image will be very bright (hyperechoic) in this part of the ultrasound image.

The time duration for receiving echoes from a specific depth equals (2× depth)/speed of sound; from a depth of 15 cm, the time will be 195 μs.

A-mode scanning (Amplitude mode scanning): In the early days of ultrasound, this was the only way possible to do the scanning. For each echo received, the amplitude of the echo was displayed on a screen, with the amplitude displayed as the *Y*-axis and the time as the *X*-axis.

B-mode scanning (Brightness scanning): In B-mode scanning, the amplitudes are converted into different gray levels, and the gray levels in the different parts of the tissue being scanned are displayed with varying gray scale levels on a map with the depth of the tissue as *Y*-axis and the position along the transducer surface as the *X*-axis.

For a B-mode image, modern equipment usually uses a gray scale resolution of 256 levels. Liquid collections like cysts and the gall bladder will appear black, while areas with many strong reflectors such as bony structures will appear echo-rich or even white.

The Propagation of Sound in Tissue

When an ultrasonic wave is moving down through the tissue, the actual movement is influenced by:

Speed of Sound

Even though the speed of sound in tissue is relatively constant (apart from being very different in air and bone with total reflection as a result), it is slightly lower in fatty tissue than in muscle or normal kidney. Usually the ultrasound system is set up to use a speed of sound of 1,540 m s^{-1}. If the actual speed is different, this has an impact on how accurately a point is displayed and on how accurate measurements are.

Attenuation

The attenuation of the sound wave is very dependent on the frequency used. It will also increase with increasing depth of the tissue. The attenuation is due to the absorption and reflection of sound energy in tissue, in particular when most of the energy is reflected due to a big difference in impedance. When the ultrasonic wave meets air or bony structures, it is almost completely reflected. As a result, a shadow artifact appears behind the area where air or bone is encountered.

This phenomenon means that for good image quality, it is important to ensure good contact (no air) between transducer and tissue. Ultrasound gel is used to avoid too much impedance difference between the transducer surface and the tissue.

For transrectal scanning of the prostate, it is essential that no air is trapped between the transducer and the rectal wall. In cases where a water-filled balloon is used, air bubbles must be eliminated from the balloon as well.

The attenuation means that it is important to use as high a frequency as possible without sacrificing penetration depth, keeping in mind the depth of the intended organ.

The operator has to use the time gain compensation (TGC) function to compensate for this attenuation in the tissue. Without this, the image will be darker and darker as you go from the transducer surface down through the tissue. All scanners have TGC to allow a depth-dependent gain adjustment in order to compensate for the tissue attenuation.

Focusing the ultrasonic beam

See Fig. 11.1.

During the use of an ultrasound transducer, the operator must try to make sure that the area to be examined is placed within the focal range. With electronic transducers an important adjustment available for the operator is the ability to move the focus as close to the area of interest as possible.

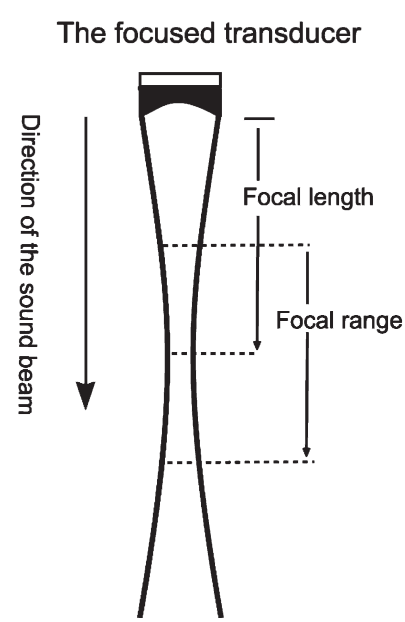

Fig. 11.1. Illustration of the form of the ultrasound beam of an electronic transducer, where the focal range indicates the range of depths giving the best resolution for the transducer. The Focal range is the distance from the transducer array to the depth, where the sharpest focus will be obtained

Limitations of Ultrasound

Resolution

Resolution plays an important part in optimizing the image quality during scanning. Resolution is divided into different categories: axial, lateral, and contrast resolution.

The axial and lateral resolutions are decisive for separation of the different reflecting structures. If the resolution is insufficient, two reflectors may be displayed as one. The minimum displayed lateral size will be equal to the beam width, and the minimum axial dimension will be equal to one-half of the pulse length.

Axial Resolution

This is the resolution in the direction of the ultrasonic beam. It depends very much on the length of the pulse used – a shorter pulse gives a higher axial resolution, but when the pulse becomes shorter, the sensitivity and penetration are reduced. The axial resolution also depends on the ultrasound frequency – higher frequencies correspond to shorter wavelengths. Two points in the tissue cannot be distinguished if they are located within one wavelength of each other.

For the operator, the important adjustments are the choice of transducer and – for a given transducer – the choice of frequency. The pulse form and length is built into the equipment and cannot be changed by the operator.

Some transducers have a fixed frequency, but in most modern equipment, different frequencies can be selected for the same transducers.

In addition to this some recent ultrasound systems have the possibility of using a high frequency in the surface and gradually reducing the frequency in the lower part of the image, thus optimizing the image for high near-field resolution, but keeping an adequate penetration depth.

See Fig. 11.2

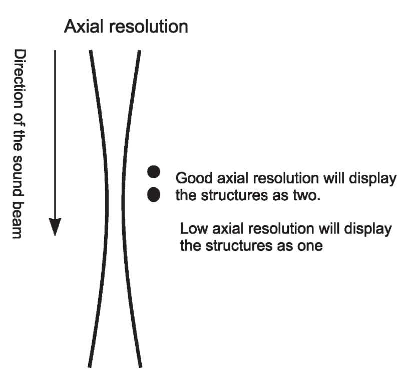

FIG. 11.2. Illustration of the effects of axial resolution. The resolution in the direction of the beam is called the axial resolution and can be adjusted by changing the frequency. Higher frequencies will offer better axial resolution

Lateral Resolution

The lateral resolution of an electronic transducer depends on the width of each ultrasonic beam and on the density of lines in the image. Optimizing the pitch of the transducer and electronic focusing of the beam are the two main ways of optimizing the beam width in the area of interest.

Pitch is a description of the size of each individual crystal in an array. A smaller crystal means a smaller pitch and a finer resolution, but sometimes at the cost of good focus in the deeper part of the tissue. Near-field transducers use high frequency with a finer pitch; abdominal transducers for kidney scanning use a lower frequency with a larger pitch. For the operator, the relevant adjustments affecting the lateral resolution are adjustment of the line density and optimizing the focus position. The sharper the focus and the higher the line density, the better the lateral resolution at that point. The cost of higher lateral resolution is usually the frame rate: in order to optimize the axial resolution, the image frame rate will be reduced (Fig. 11.3).

Contrast Resolution

This term is used as a measure of the system's ability to distinguish between two parts of the tissue with almost – but not completely – identical echogenicity: if the system can distinguish well between tissues that are very similar, the contrast resolution is high. The actual contrast resolution is an integrated characteristic of the combined ultrasound system and transducer.

All ultrasound systems allow the operator to adjust the gray scale and dynamic range. This can be important for creating a diagnostic image, in which it is possible to easily differentiate pathology and normal tissue. It should, however, not be used to compensate for an ineffective adjustment of the monitor.

The monitor must be correctly adjusted before other image adjustments are attempted.

Tissue harmonic imaging can sometimes be used to increase the contrast resolution.

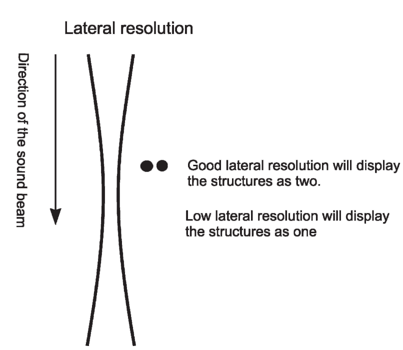

FIG. 11.3. Illustration of the effects of the lateral resolution. The resolution in the image plane perpendicular to the ultrasound beam is called the lateral resolution. In order to obtain a better lateral resolution, a higher line density or a sharper focus can be used

Artifacts in Ultrasound

In ultrasonic imaging, artifacts are areas in the image that are not indicative of the tissue being examined. They can be caused by the instrumentation or related to physical phenomena that the instrument is not able to compensate for.

Examples of artifacts often encountered during a normal scanning situation are:

- Enhancement
- Shadowing
- Reverberation
- Mirror artifacts
- Propagation speed errors

Some artifacts, like enhancement and shadowing, can be very helpful for diagnosing structures like cysts and calculi. Other artifacts, like reverberations or artifacts due to poor contact between transducer and tissue or the presence of strong reflectors, may disturb the image and reduce the diagnostic value of the ultrasound images obtained.

Propagation speed artifacts are due to basic assumptions not being fulfilled. The equipment assumes that sound in tissue travels in straight lines with a uniform velocity of 1,540 m s^{-1}, and that only echoes from the transducer axis are received.

These assumptions unfortunately are not always true.

Enhancement

Increased echogenicity from tissues behind areas with low attenuation. This type of artifact is normally seen behind cystic or other liquid collections. This kind of artifact helps in identifying cystic structures and making sure that the structure is a true cyst (Fig. 11.4).

Shadowing

Shadowing happens due to a decrease of echogenicity from tissues behind a zone with strong reflectivity or attenuation. This artifact occurs behind strongly reflecting structures like calculi or bony structures (for example, the pubic bone). A so-called acoustic shadow behind a strong reflector (for example, the bone or calculus) is the result. In order to have a closer look at the tissue located in the shadow, the transducer must be reoriented so that the shadow does not cover the desired area.

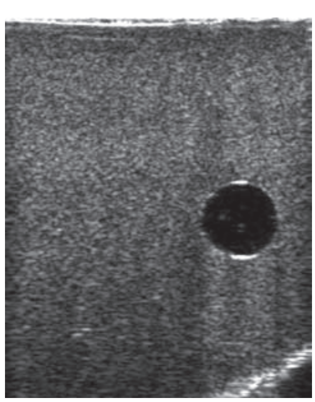

Fig. 11.4. A cystic structure showing enhancement behind the cyst, because the cyst attenuates the beam less than the surrounding tissue does

Reverberations

When two or more strong reflectors are present, multiple reflections between these reflectors and the transducer surface may occur. The reverberations are caused by internal re-reflections in the tissue, or between the transducer and a reflector in the tissue. The rectal wall and a water balloon, often used, can form reverberation artifacts if air bubbles or other materials are located between the transducer and the area to be examined (Fig. 11.5).

If the ultrasound beam does not hit an interface at a perpendicular angle, the direction of the beam will be altered. The equipment assumes straight-line propagation when it calculates the image, so a reflector may not be displayed in the correct position. These artifacts can often be avoided by trying to scan at a perpendicular angle.

One troublesome result of refraction is called the anisotropic effect, frequently seen during transrectal scanning of the prostate. Ultrasonic beams hitting the prostate near the neurovascular bundles will hit the prostatic border in a tangential manner. Therefore a significant part of the beam will be reflected in other directions than the direction of the incident ultrasound beam. As a result, a lower intensity will be received by the transducer, and, due to attenuation, the echoes from these areas will be displayed as darker areas. This could be mistaken for suspicious hypoechogenic areas, but is just a result of the attenuation due to hitting these areas in a tangential manner (Fig. 11.6).

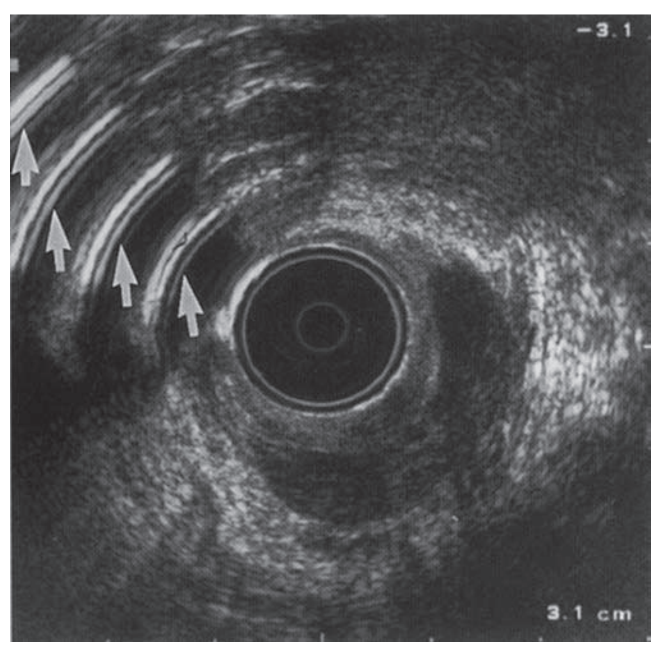

Fig. 11.5. Strong reflection-artifacts (*arrows*) due to air trapped between a water-filled balloon and the rectal wall

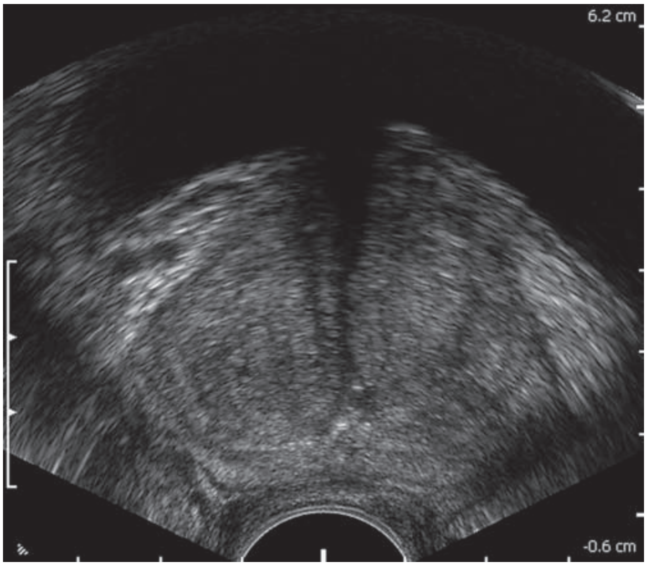

FIG. 11.6. Anisotropic effect demonstrated at the left and right-hand corner of the prostate

Mirror Artifacts

If the ultrasound beam hits a strong reflector, a mirror image of a real structure is seen on the other side of the reflector. This artifact can usually be avoided by changing the position of the transducer.

When scanning the liver and kidneys, this artifact is often seen when the diaphragm is hit at an angle close to 90°.

Propagation Speed Error

The sound speed in tissue is assumed to be 1,540 m s^{-1}. if the actual speed is higher or lower than this, a structure at a certain distance will be displayed as being closer or further from the transducer than it really is.

Newer Techniques and Methods in Urologic Ultrasound

Prostate Harmonic Imaging

When an ultrasound wave passes through tissue, it becomes distorted, and additional frequencies that were not present in the fundamental signal are generated. Multiples of the fundamental frequency are called harmonics, and the second harmonic frequency is of particular interest. The use of this feature – a scanning modality called tissue harmonic imaging – enhances the visibility and detection of hypoechoic structures. It also seems to suppress some of the detrimental effects of the presence of hyperechoic structures.

Propagation of ultrasound in any medium is determined by the Impedance (Z = the density of the medium multiplied by the velocity of ultrasound in that medium) and by the Reflection Coefficient ($r = Z/SZ$). The ultrasound wave becomes distorted as the tissue expands and compresses in response to the wave. This nonlinear distortion results in the generation of additional frequencies not present in the original waveform.

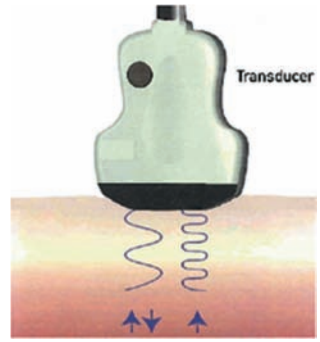

FIG. 11.7. Fundamental and second harmonic frequency

The reflected signal thus not only includes the fundamental frequency but also multiples of this frequency. The frequency that is double the fundamental frequency is called the second harmonic.

The second harmonic frequency for imaging has been used for some time for scanning the liver and kidneys but has now also become an imaging possibility when scanning the prostate.

Harmonic imaging of the prostate combined with random plus targeted biopsies may prove to be advantageous in increasing the sensitivity of transrectal prostate ultrasound (Fig. 11.7).

Grating Lobes

In ultrasound it is assumed that all energy is transmitted from the transducer in the expected direction of the ultrasound beam. Unfortunately this is not true. The main part of the energy is transmitted in this manner – this is called the main lobe. Part of the energy is, however, transmitted in other directions – called the side lobes. Energy falling outside the main lobe in the sound beam from an array transducer is a result of the active transducer aperture being split into elements. This phenomenon is called Grating Lobes (or Side Lobes). The energy in these lobes is substantially less than the energy in the central ultrasound beam (the main lobe) but is inversely proportional to the radius of curvature of an array probe.

A prostate ultrasound probe will always tend to have a small radius of curvature in order to keep the outer diameter of the probe a reasonable size, and therefore side lobes can sometimes create artifacts and degrade the image quality.

Grating lobes are particularly disturbing in prostate ultrasound. The lobes will extend almost laterally out from the probe because of the small radius of curvature. When passing through the periprostatic fat tissue, the grating lobes will eventually hit the inferior side of the pelvic bone, where the difference in impedance to the ultrasound is very high. This shift in impedance will cause an almost 100% reflection of the energy in the grating lobes, hence the energy will bounce back across the prostate ultrasound image, resulting in a degradation of the image quality.

Strong second harmonic signals are generated in a region of high sound pressure, and accordingly only weak signals are found in the region where the grating lobes are being generated. But with almost total reflection, the lobes are disturbing enough.

Using the second harmonic imaging technique not only reduces the effect of the lateral grating lobes but also reduces the angle of the second harmonic lobes. The result is that the lobes are more parallel to the sound beam. Hence the risk of the lobes being reflected from the pelvic structures is also minimized (Fig. 11.8).

The true advantage of prostate harmonic imaging may be the enhancement of any hypoechoic structures combined with

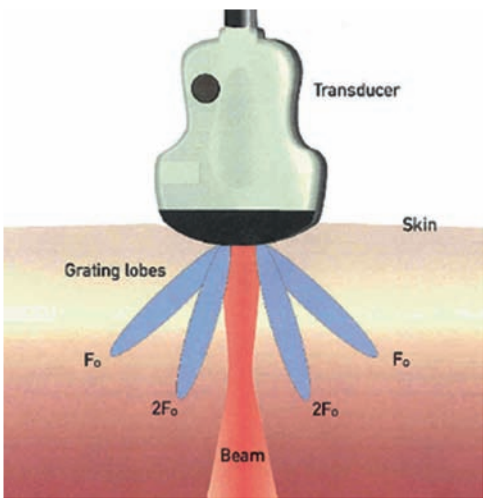

Fig. 11.8. Grating Lobes. F_o = Fundamental Frequency, $2F_o$ = Second Harmonic Frequency

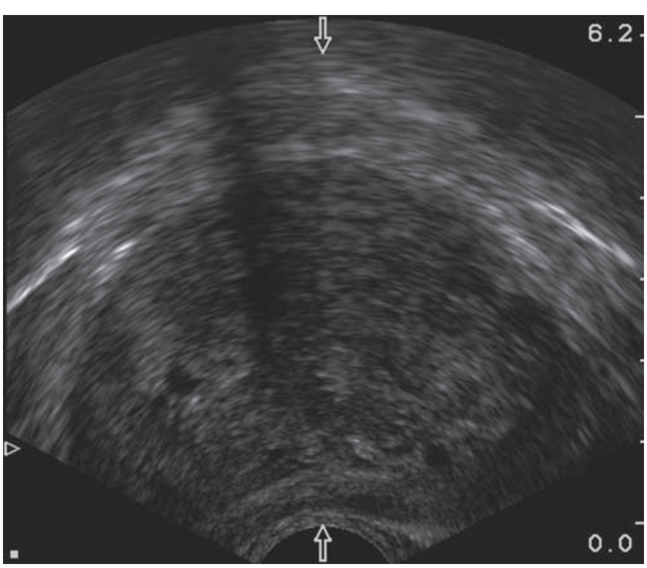

Fig. 11.9. Conventional ultrasound image illustrates a 77-cm^3 prostate without any focal lesion or nodules

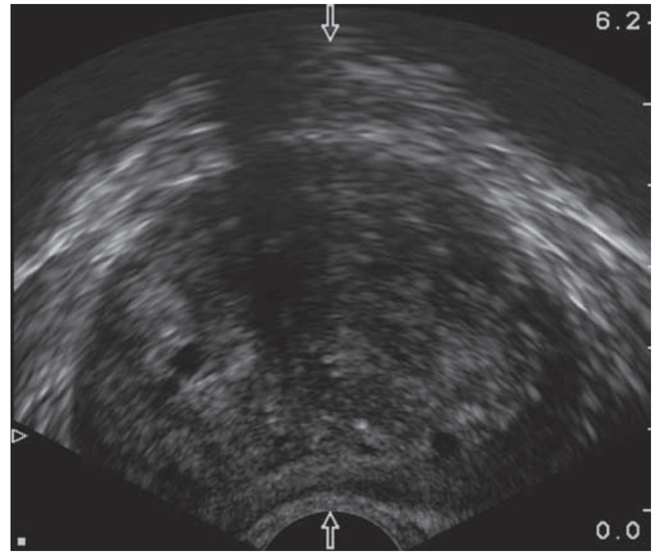

Fig. 11.10. Same patient scanned using harmonic imaging. Note the increased resolution in differentiating between the peripheral zone and the transitional zone

the suppression of hyperechoic phenomena, such as shadowing due to the ultrasound beam being reflected from *corpora amilacia* (Figs. 11.9 and 11.10).

Compared to conventional TRUS, tissue harmonic zoography allows better visualization of malignant lesions resulting in improved differentiation and better detection of small prostate masses. It appears to be a promising tool for improving the diagnostic yield of prostate TRUS.

The transitional zone and the rather compressed peripheral zone are often better seen on the harmonic imaging picture because of reduced grating lobes.

3D Ultrasound

3D ultrasound has been employed in different clinical applications for several years. The acquisition of a 3D data set and the techniques employed are, however, not the same in different applications. The most commonly known version of render mode is *surface render mode*, used extensively to produce early images of the face of a fetus.

Surface render mode only gives good results when a surface is available to render. These techniques fail when a strong surface (a shift in the ultrasound impedance of tissue) cannot be found, as is the case in the subtly layered structures within the anal canal, rectal wall, prostate, etc.

High-resolution 3D ultrasound acquires four to five transverse images per mm of acquisition length. Because of this high resolution, which typically is close to – or equal to – the axial and transverse resolution of the 2D image, 3D postprocessing facilities can reveal significantly more features than can be seen in relatively low-resolution 3D data sets obtained in other applications. Combined with the *volume render mode*, 3D scanning has the potential to give higher spatial resolution for assessing prostatic disease, compared to what is possible using 2D ultrasound.

A 2D ultrasound image has under normal circumstances almost no depth, because of the requirements of keeping the depth of the image as small as possible.

Volume rendering mode techniques use what is called a ray tracing model as the basis of operation. A ray or beam is projected from each point on the viewing screen (the display) back into and through the volume data. As the ray passes through the volume data it reaches the different elements (voxels) in the data set thereby creating the volume render view, making it possible to look deeper into the data set.

This *volume render* effect may in particular be dramatic if a number of vowels inside an acquired 3D data set are produced from scanning hypo echoic structures. A good example is the use of this technique to assess hypo echoic prostate lesions. Voxel values behind, for example, a strongly reflective interface will also result in the illusion of looking into a semitransparent dark cavity in the anatomy.

It is also possible to apply other render mode projections:

Maximum Intensity Projection (MIP) tries to find the brightest or most significant color or intensity along a ray path.

Transparent modes allow the separation of color and intensity data and selective control of the transparency of the two components. Using this method, it is possible to reduce the intensity of the gray scale voxels so that they appear as a light fog over the color information. Color information hidden behind an obstruction can then be made visible.

Furthermore, and possibly more important, acquired 3D data sets open up for completely new postprocessing techniques where, for example, the data block can be made opaque, resulting in additional depth information.

3D acquisitions give access to a new, wider range of viewing planes that are unavailable with 2D scanning. 3D ultrasonography reconstructs a "volume" from 2D scans. This volume can be analyzed using viewing in scan planes not accessible by 2D ultrasound – for example, the new coronal view of the prostate. 3D ultrasonography permits quantitative examinations and excellent measurement capabilities.

A major advantage of working with the 3D system is that images used for diagnosis are totally reproducible. Using the original electronic data, images can be reviewed as many times as needed.

The Specific Benefits of 3D Ultrasonography of the Prostate

3D ultrasonography enables a simultaneous view of the sagittal, transverse, and – particularly important – coronal planes. By providing such variety of different views of the prostate, urologists get an invaluable tool for diagnostics with 3D imaging. They can more easily distinguish the zones of the prostate and see small lesions.

3D prostate imaging makes spatial relationships much clearer, so urologists can better assess the extent of disease. It also provides information that can be valuable for selecting patients for alternative therapies, for example external beam radiation or prostatectomy/radioactive seed implantation. In particular, it may be easier to determine whether the "prostatic capsule" has been penetrated, a key factor in tumor staging. The 3D data set is acquired in a precise, controlled manner, and can be evaluated at any depth, plane, or angle. It can easily be saved for interpretation off-line.

The 3D acquisition is performed using a precision motorized device, and a much higher number of images are interpreted for the total image reconstruction (typically several hundred). This technique has resulted in a much improved spatial resolution.

The 3D volume is constructed from the sequence of acquired images using interactive 3D software built into a 2D ultrasound system. The software allows the 3D image to be sliced in any orientation as well as rotated in any direction. The display software allows up to three surfaces of the prostate to be viewed simultaneously in a 3D volume rendering. Visualization of lesions in three planes appears to allow improved assessment of extracapsular extension. The 3D image is formed with the coronal in addition to the standard sagittal and axial planes, which means that instead of a single transverse or sagittal scan section of the prostate, as seen with 2D transrectal ultrasonography, the 3D images provide an added surface to be visualized. This surface could be the lateral, posterior, anterior, superior, or inferior surface of the prostate, or a through section of the gland where the posterior surface of the prostate shows tumor infiltration beyond the periprostatic fat.

Examples of 3D Ultrasonography of the Prostate

See Figs. 11.11 and 11.12

3D views of a prostatic carcinoma. By manipulating the data, views can be obtained, where capsular penetration is demonstrated. By means of 3D manipulation of the data obtained views

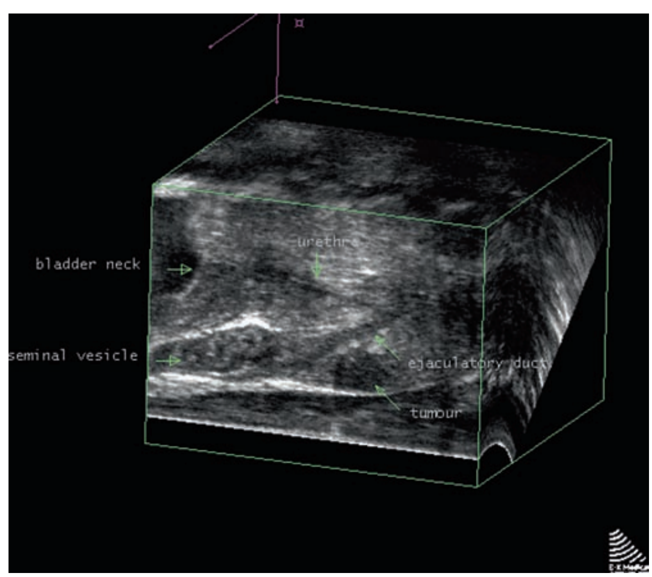

Fig. 11.11. Adjusting the plane and viewing angle of the 3D datas proved the lesion to be penetrating the prostate capsule a fact not seen during the 2D prostate scan

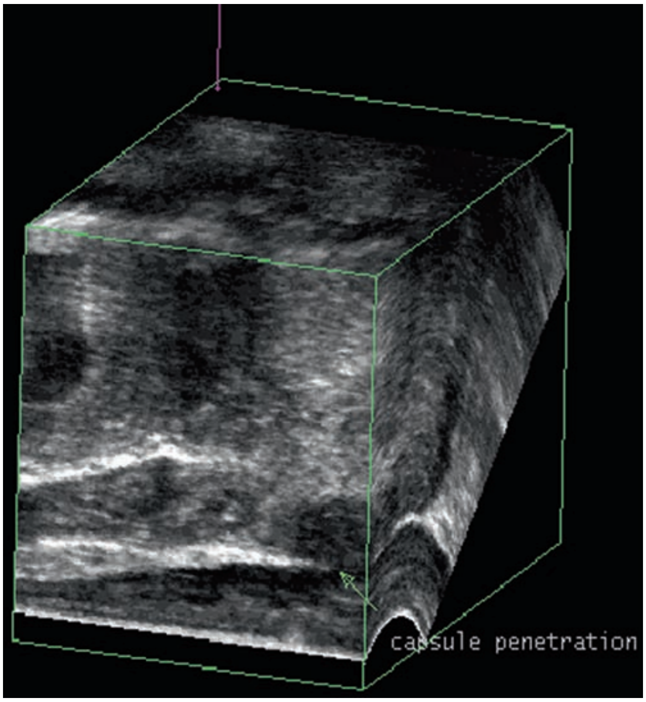

FIG. 11.12. From the 3D image of the prostate a small lesion of the peripheral zone seems not to be penetrating the capsule. This was also the conclusion from the prior 2D prostate scan

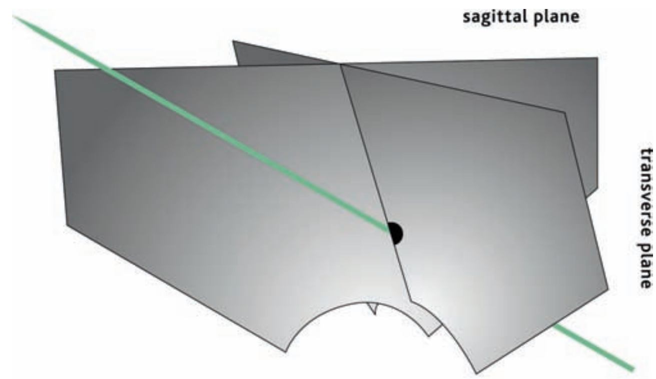

FIG. 11.13. The two simultaneous image planes of the simultaneous biplane transducer. Both image planes are projected nearly perpendicular to the longitudinal axis of the transducer. The transducer is manufactured with a forward tilt of both arrays for better patient tolerance, as the depth of introduction into the *ampulla recti* is correspondingly less

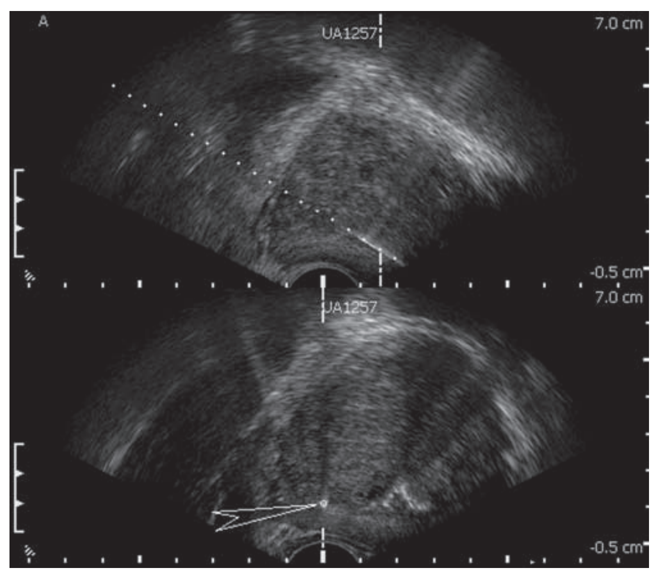

FIG. 11.14. Targeting the right lateral peripheral zone of the prostate ultrasonically guided precision biopsy. The *arrow* on the transverse (*lower*) image points to the marker that indicates the projection of the needle path in the perpendicularly opposite image plane, the sagittal (*upper*) image. Note that this target is exactly in the lateral peripheral zone. Many urologists consider biopsies of this area difficult if they are only based on the sagittal image

not possible to obtain by 2D scannings, e.g., coronal views can be used to obtain additional diagnostic information.

Recent Developments in Transducer-Design for TRUS

Until recently, the preferred method for performing TRUS was either to use an endfire transducer for prostate scanning, or to use the biplanar approach for the scanning. Different scan planes can be visualized using both methods, but until a few years ago none of the scanning methods could visualize more than one scan plane at a time.

A one-plane view of the prostate can make it difficult to carry out a precise biopsy regimen because you can't be certain that you are sampling from the intended targets. This applies particularly when you are attempting to use a strategy of placing some biopsies in very lateral positions. Some recent scientific publications have stated that such a strategy seems to lead to better efficiency for detecting the cancer.

Particularly in patients with the peripheral zone compressed due to BPH, it can be very difficult to be certain that the needle is hitting the intended target in the lateral part of the peripheral zone.

A few years ago a new kind of TRUS transducer made it possible to scan in simultaneous biplane, i.e., scan in transverse and sagittal planes at the same time. Based on the isocenter idea – that the design must make the needle echo visible in both planes in one point (the isocenter) – this concept makes it possible to make sure of the exact placement of the needles (Figs. 11.13–11.15).

The simultaneous biplane view makes it possible to target the areas you intend for biopsies more correctly and with great precision. The combined transverse and sagittal views make it possible for you to see exactly where and how deeply your biopsy needles are placed.

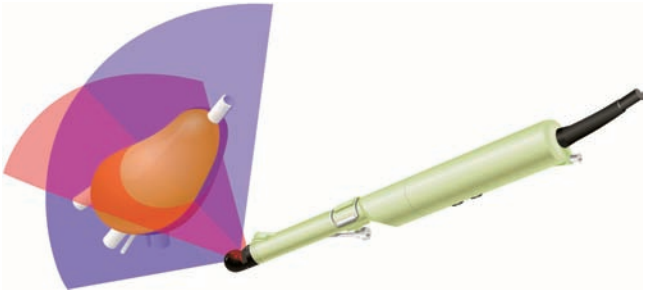

FIG. 11.15. An illustration of the two simultaneous scan planes for scanning the prostate

FIG. 11.16. Transverse image of large prostate

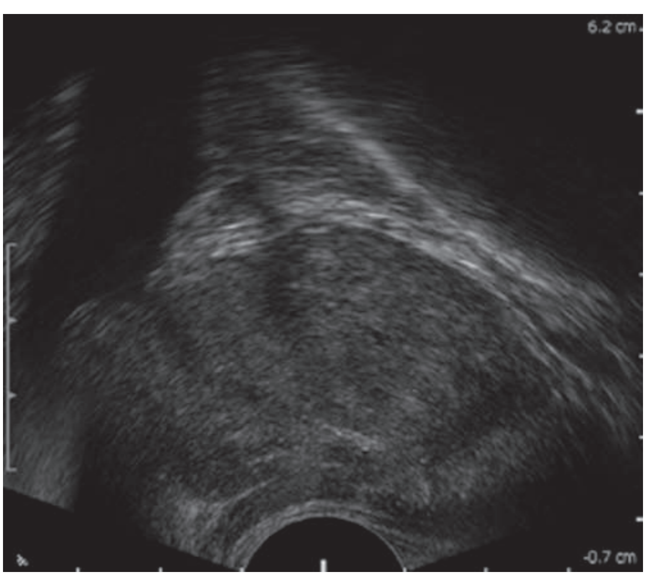

FIG. 11.17. Sagittal image of a large prostate

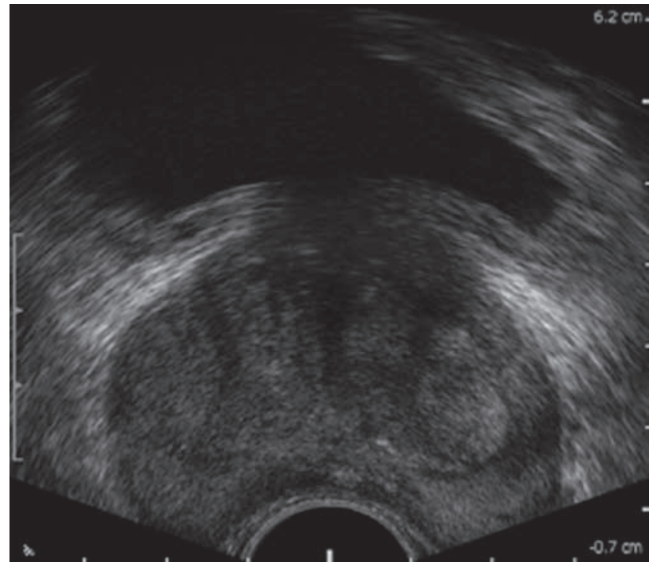

FIG. 11.18. Endfire image of a large prostate

Simultaneous biplane is particularly valuable for targeting more lateral biopsies of the peripheral zone with greater precision because the peripheral zone, certainly in elderly patients with BPH, becomes very thin and difficult to target.

A real-time transverse image together with a real-time sagittal one provides a clear indication that the needle is in the intended area, for quicker and more accurate biopsies.

In some patients, however, simultaneous biplane has some limitations for biopsies in the apical part of the prostate. The biopsy regimen should include biopsies that are from the left and right apical part of the prostate, yet still close to the parasagittal plane.

Endfire imaging, although offering less confidence for the precise placement of the lateral biopsies, is ideal for taking apical biopsies, because the biopsy guide for endfire imaging is placed immediately behind the imaging array.

Thus the ideal transducer for biopsies close to the prostate apex seems to be an endfire type. The ideal transducer for placing lateral peripheral zone biopsies is the simultaneous biplane transducer. By combining the two concepts: real-time simultaneous imaging and the biopsy route through the midcentral transducer finger – and the endfire guidance of the biopsy route, the biopsy technique remains optimal for all individual biopsies of the prostate.

One transducer, using one dual biopsy guide and one insertion, combines the best from endfire scanning with simultaneous biplane scanning.

This scan technique for TRUS combines simultaneous biplane scanning and biopsy with endfire imaging and biopsy in the same transducer. One insertion of the transducer combines the best from endfire scanning with the best from simultaneous biplane scanning.

This new technical development makes it possible to take biopsies in all sections of the prostate with excellent orientation and confidence (Figs. 11.16–11.19).

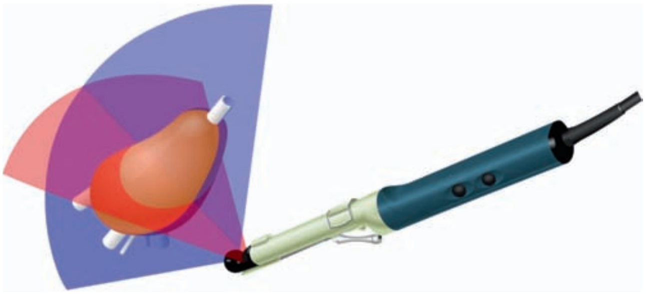

Fig. 11.19. Illustration of the two scan planes when the transducer is used in simultaneous biplane mode

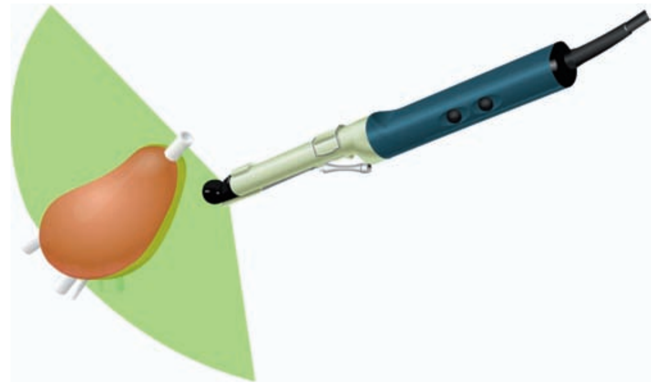

Fig. 11.20. Illustration of the scan plane when the transducer is used in endfire mode