Ultrasound Image Optimization ("Knobology"): B-Mode

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ABSTRACT

Ultrasound is a ubiquitous and indispensable diagnostic and therapeutic tool in medicine. Due to modern equipment and automatic image optimization, the introduction of ultrasound imaging currently requires only little technical and physical knowledge. However, in-depth knowledge of the device functions and underlying mechanisms is essential for optimal image adjustment and documentation. From a medical as well as an aesthetic point of view, the goal should always be to achieve the best possible image quality. The first part of this article provides an overview of the handling of ultrasound systems, fundamental adjustments, and their optimization in B-mode ultrasound.

Introduction

"Knobology" describes the pertinent knowledge and use of ultrasound (US) equipment to achieve the best settings and applications for patient care. US does not require ionizing radiation and is an indispensable imaging method in medical diagnostics. A major limitation is that ultrasonic waves are transmitted neither through bone nor air, which restricts its use. Apart from good knowledge of anatomical structures and examination techniques, knowledge of how to achieve the correct machine adjustments for the best image quality and to maximize the potential of US equipment functions is essential [1]. In the following document, we present techniques to optimize general device settings and to achieve optimal use of B-mode.

* Both authors contributed equally to this work.

How to Manually Boot, On/Off

The US system is powered on and off using the partial power on/off control located on the control panel. A steady color indicates that the power is on. In some systems, flashing color indicates that it is plugged into the power supply and the circuit breaker on the US system is in the ON position. No color indicates that the power is off, unplugged from a power supply, or the circuit breaker on the US system is in the OFF position. Modern mobile and handheld US systems are equipped with a battery, which is automatically charged when the system is on power supply. Due to complex software functions, booting and shutting down the system may take some time. To avoid damage or disturbances of the US system, the power supply should not be switched off while the US system is in the process of booting up or shutting down.

Monitor

The monitor presents the information generated by the US exam and makes it available for the user to review. The examination room should be darkened as much as possible in order to avoid a loss of contrast on the screen due to incident light. It should also be considered that the human eye needs about 20–30 minutes to achieve complete dark adaptation. If the room is entered or darkened shortly before the examination, the viewer of the US image could miss details that can only be perceived after dark adaptation [2, 3]. The position of the monitor should allow for a near-perpendicular viewing angle for the examiner. An overly flat angle leads to a loss of contrast perception. Finally, the display technology has to be considered: A liquid-crystal display (LCD) is a flat-panel display thought superior to the cathode-ray tube (CRT) monitors used in the past [4]. The use of organic light emitting diodes (OLED) is a newer flat light-emitting technology that provides increased contrast, but is still very expensive and currently only installed in a few devices [5]. Some mobile and handheld US devices are equipped with touchpad monitors, which allow partial or complete management of system functions.

The Right Transducer

Crystals located at the tip of the transducer are agitated by an electrical current and generate ultrasonic waves, which are transmitted into tissue via a coupling gel between the probe and the patient's skin. This process is based on the principle of the piezoelectric effect [6]. The same crystals serve as receivers of returning ultrasonic waves, which means that soundwaves reflected back from the tissue are absorbed by the transducer crystals and then generate specific electrical signals. These signals are decoded and processed into imaging information. While millions of soundwaves per second are produced when the crystal is agitated by a current and functions as a transmitter, over 99 % of the time is allocated to receiving the returning soundwaves. This assures that all returning echoes are registered, because it takes longer to receive returning waves from greater depths. The echoes of the emitted beam are usually scattered and reflected when encountering inhomogeneous tissue or impedance jumps at the transition between two tissue types or structures. This will create returning waves with different strengths and transit times. This information is evaluated by the US machine software and results in the image displayed on the monitor [7].

Three different transducer types are commonly available. They complement each other and serve different requirements with regard to body region, structural representation, depth penetration and field of view. This is made possible by variances in geometry, crystal arrangement and crystal activation.

Convex transducers (curved linear array) are typically used in abdominal and pelvic sonography. Crystals are arranged next to each other along a curved (convex) surface. This enables a widened field of view, especially in the depth display, but also assures good near field resolution. The resulting image is cone-shaped with the diameter increasing with the depth. Curved transducers with a small aperture and wide scanning field are available for transcutaneous, intraoperative, and intracavitary use. Variation of the width of the scanning field is possible with modern convex transducers.

Linear transducers (linear array) use crystals positioned next to each other in a straight line. Thus, the ultrasonic waves are arranged in parallel and produce a higher and more uniform resolution at the expense of depth penetration as high frequencies are used. The output image is rectangular.

Vector array transducers are a variety of classic linear array transducers that enable trapezoidal widening of the acoustic window. This worsens the resolution at a depth, but expands the width of the image defined by the aperture of the transducer. The technology allows for the depiction of extended structures and is used to evaluate superficial structures with high resolution such as the thyroid gland, superficial vessels, intestines, soft tissues, and joints. However, it produces more artifacts when applied to curved parts of the body [7]. An additional function of vector array transducers is image steering, which denotes lateral canting of the US window by up to 30 degrees. This may be useful to examine superficial structures which otherwise would be hidden in the acoustic shadow of totally reflecting structures (e. g. ribs).

Sector transducers (phased array) use smaller and narrower individual crystal elements that are arranged in a horizontal or circular pattern and have a smaller footprint. The functional difference lies in the control of the individual crystal sections. Through a slight time and phase offset, spherical sound fields are generated, which result in a fan- or pie-shaped image. Thus, this type of transducer is more effective than the convex transducer for depth display but loses a lot of information in the near field. The dimensions of the transducer facilitate its use for narrow acoustic windows as they occur in intracavitary sonography, echocardiography and neurosonography [8]. An important advantage of sector transducers is the application of continuous wave (cw) Doppler US, which is not possible using the other types of US transducers.

In addition to the conventional transducers that are connected to the US machine via a cord, the use of newer wireless probes might become more prominent, allowing for a more comfortable exam, especially in the field of interventional imaging. Furthermore, smaller and more mobile US devices have been developed [9, 10].

Image Quality

Image quality depends on several factors. Above all, the transducer must be adequately coupled to the patient with a sufficient amount of coupling gel. This avoids interposition of air between the transducer and the skin and ensures that all crystals are able to transmit and receive soundwaves. To achieve optimal image quality, adequate depth penetration, image width, spatial and temporal resolution, image contrast, artifact suppression [11, 12], and application of zoom are relevant. [7, 13] Hence, the best possible image parameter settings should be achieved. The above-mentioned factors will be discussed in more detail in the following sections. The goal should be to generate an image that is as realistic and aesthetic as possible, which is representative of the depicted anatomy and provides a meaningful clinical contribution.

Depth Penetration

Depth penetration influences the size of the examination window and the reproduction scale. It depends on transducer frequency, transmission power, and Tissue Harmonic Imaging (THI). Several aspects should be considered when determining depth penetration. A high depth penetration is indispensable to achieve an overview of the anatomy but goes hand in hand with slower image acquisition because an echo signal must be sent/received for each additional image line. The tissue of the patient is an important factor influencing this process since the speed of sound depends on tissue density. In addition, the layer thickness rises with increasing depth penetration, which further deteriorates the resolution. Lastly, the signal-to-noise ratio should be mentioned. This ratio increases with decreasing transmission frequency and makes it more difficult to distinguish an actual ultrasonic signal from background noise artifacts. Therefore, a lower depth penetration improves the representation of moving structures as well as smooth image reproduction during transducer movements. For these reasons, the selected depth should be deeper than the structures of interest (▶ **Fig. 1**). Finally, a more lateral insertion of the sound window (B-image angle) leads to a higher temporal resolution.

Zoom

Zoom can help to enlarge an image section on the monitor. The read-write zoom has no influence on depth penetration. It depends on the configuration of the transducer used, and thus on the maximum possible range of the ultrasonic waves. If, however, a section of the image is zoomed into, the echoes outside the remaining examination area now no longer have to be evaluated, thus improving temporal resolution as well as line density. It should be noted that by comparison, a simple magnification of a frozen image does not gain any advantages, as in digital image processing.

Resolution

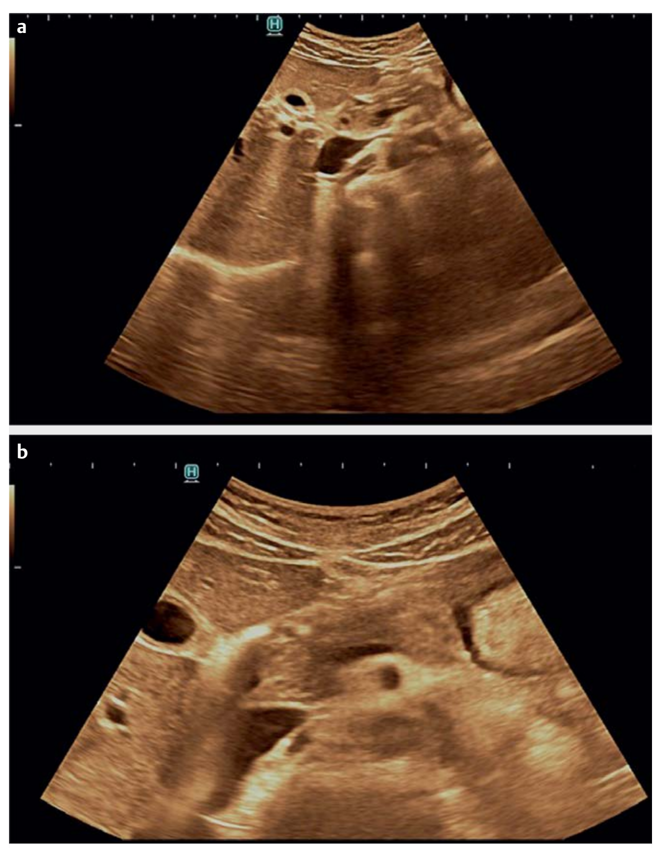
Spatial resolution

It is essential to distinguish between different terms used for image resolution, as they describe completely different aspects of US.

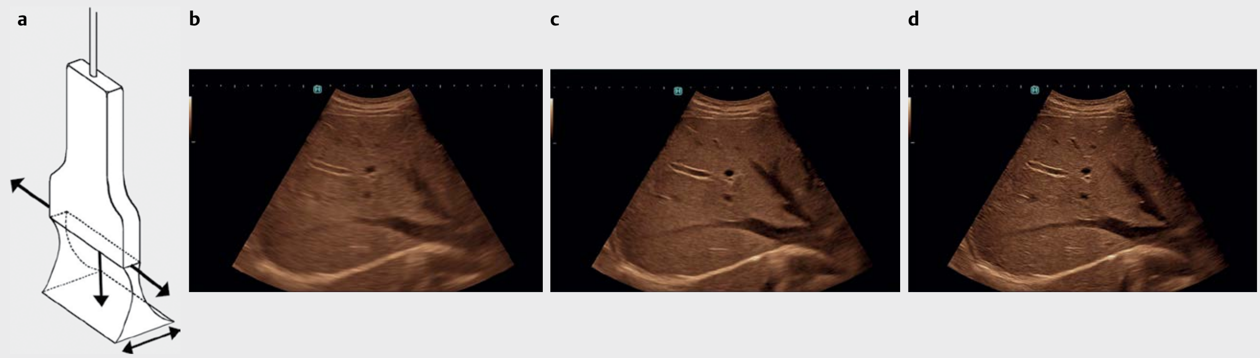
Ultrasonic waves propagate in the extended axis of the transducer, which is therefore also referred to as the axial direction. The axial resolution (▶ **Fig. 2**), depends on the nominal transducer frequency and pulse length, which both determine the axial distance at which two individual points can still be distinguished.

The second plane, called lateral resolution, defines the discrimination of two points perpendicular to the beam propagation (▶ **Fig. 2**). It is influenced by beam width and the transducer's line density. The latter is the number of pulses (scan lines) being laterally transduced. The examiner can adjust the line density to affect the resolution, but the maximum value is hardware-limited. A lower line density produces a smoother image and might be advantageous for vascular delineation, musculoskeletal tissue definition, or structures with curved or irregular borders (a stronger reflection occurs with a perpendicular angle of incidence). It should be further noted that ultrasonic signals must be transmitted and received for thinner and thus more lines, which inevitably decreases the frame rate (temporal resolution). Lateral resolution may be improved by narrowing the acoustic window of convex transducers. Due to its manual widening ("wide view", "trapezoid image") the anatomical overview improves at the cost of lateral resolution. Some US systems also offer a "speed of sound correction" to improve lateral resolution depending on the particular characteristics of the insonated tissue (e. g. breast vs. vessels) [14].

Although a two-dimensional image is displayed on the screen, it represents the projection of a three-dimensional structure (▶ **Fig. 2a**). Therefore, the user must also consider how much information of the Z-axis is summed up to an "infinitely" flat image. In ultra-



▶ **Fig. 1** US examination of the pancreas with inadequate **a** and adequate **b** selection of depth penetration. The field of interest (pancreas) is in the middle to the lower third of the adequate image b) but in the upper third of the inadequate image example **a**.



▶ **Fig. 2** Overview of ultrasonic axes **a** and examples for varying line density. Examination of the right liver lobe with line density at 1/8 **b**, 4/8 **c** and 8/8 **d**.

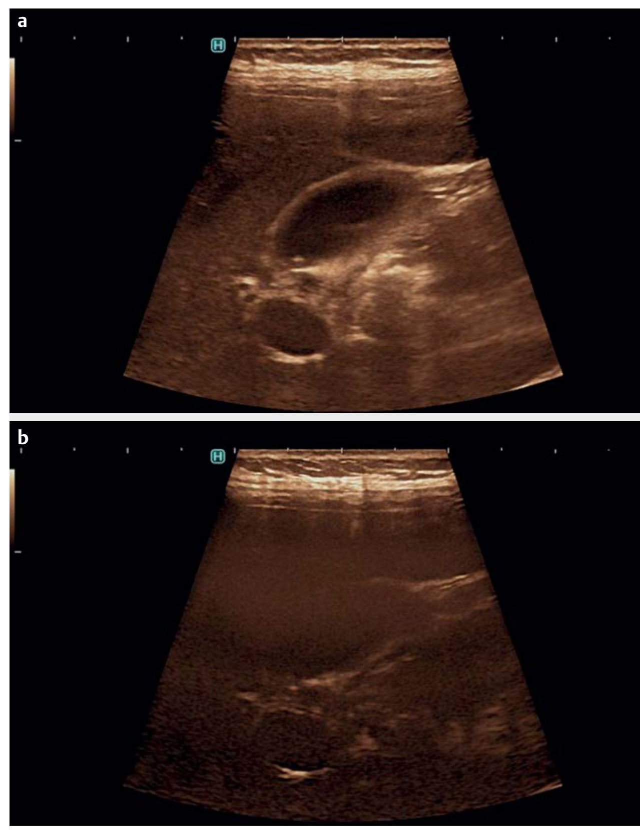
sonography, this plane is called layer thickness and depends on the selected transducer, image section, focus, and examination mode used (e. g. B-mode, Doppler, or contrast-enhanced US). The layer thickness in Doppler or contrast-enhanced US is higher than in B-mode. At low frequencies/high depth penetration, the layer thickness can be up to several centimeters and thus might create summation artifacts. Therefore, it is crucial to be aware of the difference between layer thickness and line density.

Transducer frequency

The transducer frequency is calculated from the ratio of speed and wavelength. Since higher frequencies have a shorter wavelength and the wavelength is reciprocally proportional to the resolution, the highest possible frequency should always be selected ("try high"). This setting is limited by the decreasing depth penetration that results from an increase in frequency (▶ **Fig. 3**). The nominal frequencies of the transducers are between 1 and 25 MHz and are generally determined by the transducer itself. Thus, convex transducers are usually working at 2–8 MHz. Depending on the device, either a subrange (e. g. low, medium, high) or a defined center frequency (e. g. resolution mode vs. penetration mode or the center frequency number) can be set within this range to influence depth penetration and resolution.

Tissue harmonic imaging

Tissue Harmonic Imaging (THI) improves the contrast ratio through increased lateral resolution and reduced background noise by promoting the reduction of side lobe artifacts [11] (▶ **Fig. 4**). These are lateral incident artifact echoes, which are mostly missing so-called harmonics. Harmonics are partials, which are caused by a distortion of transmitted sound waves and arise in the area of highest pressure, i. e. in the central axis area of the transmitting lobe. THI only registers ultrasonic waves that contain harmonics and thus simultaneously filters out the side lobe artifacts. In the case of "Second Harmonic Imaging", the second harmonic wave is used, which corresponds to twice the underlying transmission frequency [15–17]. Disadvantages of the past are no longer relevant today due to technology improvements, which is why it is typically not practical to switch off THI [18].

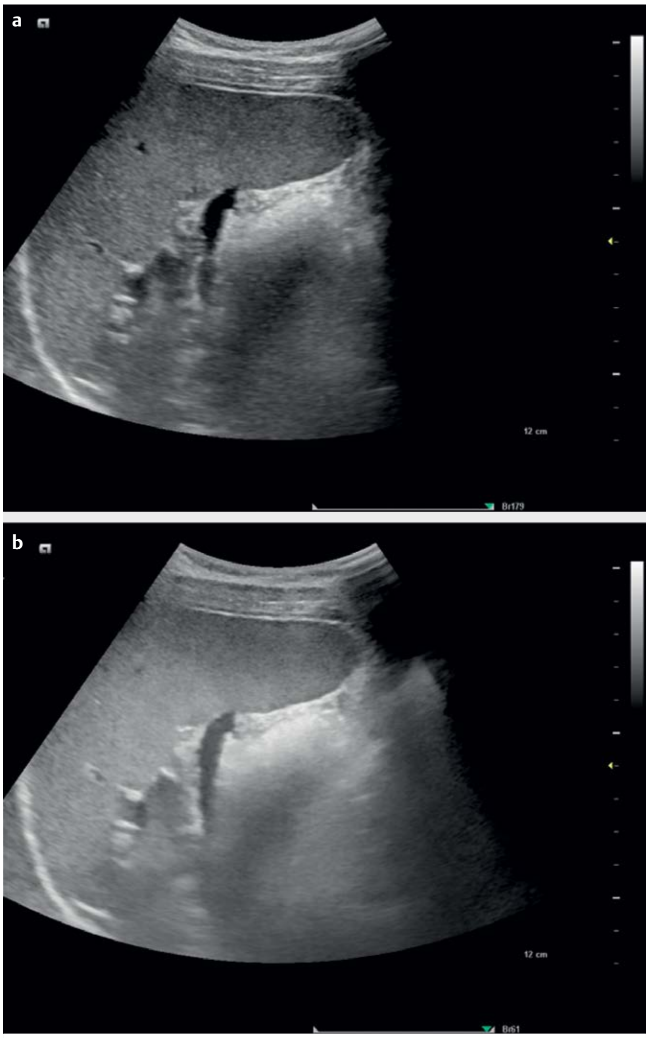


▶ **Fig. 3** Examination of the gallbladder and hepatic hilum using a linear transducer in a slim person: High spatial resolution and adequate depth penetration at lowest possible frequency **a** and loss of depth penetration using a (too) high frequency **b**.

Pulse inversion is a technique to compensate the limited bandwidths of THI. Following the normal ultrasonic beam, an inverted replica is sent and the received signals are analyzed. Pulse inversion works at all received frequencies and is therefore able to improve the resolution [13].

Compounding

Compounding is a tool for image optimization that combines multiple images resulting from multiple aperture positions (spatial compounding) or multiple transmission frequencies (frequency compounding) into a single composite frame in real time. This technique suppresses background noise, speckles, and artifacts [11] and improves resolution and contour display. At the same time, the image looks greasy and diagnostically valuable enhancements as well as shadowing artifacts may be lost. Thus, compounding is reasonable only in a limited and targeted way. In addition, it should be used according to subjective preference. Compounding consists of three different types: Spatial compounding can be achieved using several techniques, including combination of different insonation angles (beam steering), transducer rotation, and varying transducer positions . Frequency compounding can be achieved by using multiple sources at varying frequencies, or by taking several images at different frequency sub-bands [19–22]. In strain compounding, multiple strains are created by external forces inducing tissue motion [23]. The signal-to-noise ratio can be improved through processing.



► **Fig. 4** THI improves the contrast ratio through increased lateral resolution and reduced background noise by promoting the reduction of side lobe artifacts. The spleen is shown with **a** and without THI **b**.

Focal zone

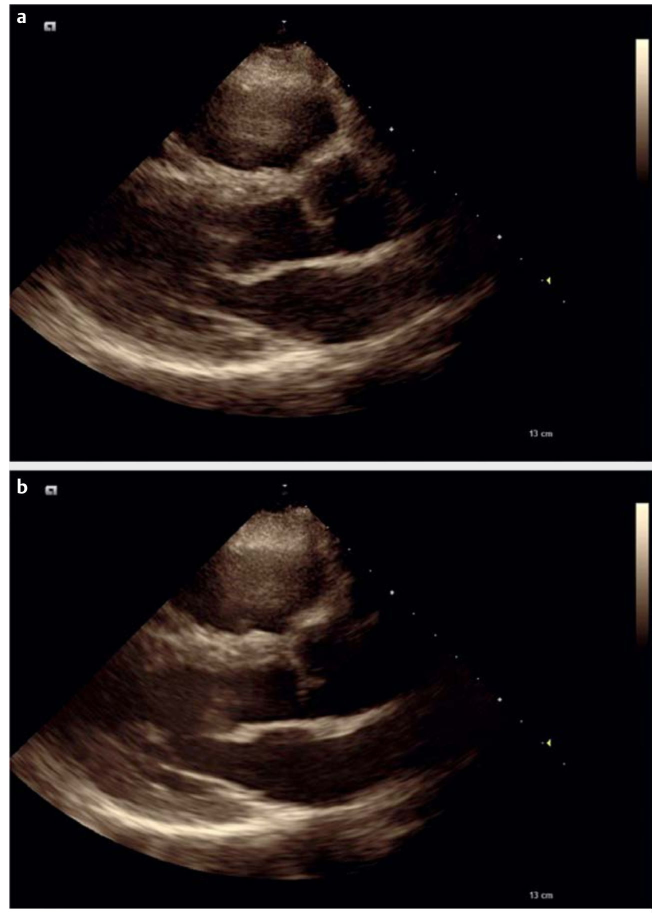
The focal zone or plane is the part of the US beam where its diameter is focused and narrowest. Proximal to this zone, the beam diameter is wider. It increases again when travelling past this zone. This causes a thicker layer width and decreased image resolution. Since the layer thickness is lowest in the focus area, it should always be at the level of interest or start just above the structure to be displayed. In the case of a cyst, the transition from the parenchyma to the fluid space to be visualized is located at the beginning of the focus area. One could argue that the echoes exiting the cyst also pass through an impedance jump required for the diagnosis. However, the principle applies that the layer thickness increases with increasing distance to the transducer, even if the focus zone is set. Classically, US devices offer one focal plane, while newer machines allow two focal planes or even a continuous focus (sometimes called eFocus or range focus) (► **Fig. 5**). An increasing number of foci lowers the frame rate.

Sectoral width

Reducing the sectoral width will improve lateral resolution by narrowing the acoustic window and can be altered when using convex probes. In most scanners, this can be easily performed by pressing the "select" button and using the trackball to alter the image width. Otherwise, you may find this on a separate button either on the console or touch screen. An example of different sectoral widths is shown in ► **Fig. 6**.

► **Fig. 5** Classically, US devices offer one focal plane **a, b**. Newer machines allow two or more focal planes or even a continuous focus (sometimes called eFocus) **c**. Examination of the pancreas using a near focus zone **a**, a far focus zone **b**, and eFocus **c**.

▶ **Fig. 6** US examination of the liver with wide **a** and narrow **b** sectoral width.



▶ **Fig. 7** Echocardiography using low persistence **a** and high persistence **b**.

Temporal resolution

In addition to the spatial resolution, temporal resolution also has a considerable impact on image quality. Influencing factors are frames per second, line density, image section, zoom and persistence (persist function). It should be mentioned that with newer devices, image section and zoom are no longer of great relevance due to much higher computing power. Accordingly, a high temporal resolution can be achieved even with large image sections. Frame rate (frames per second, FPS) must be adapted to the structure of interest. For example, a low frame rate used with convex transducers in the abdomen may be sufficient but would produce very blurred images if used with sector transducers in echocardiography (▶ **Fig. 7**).

Persistence defines how much of the previous image is taken over into the current frame. This makes the resulting live image appear smoother and less wobbly. This can be useful to a certain extent, especially when working with the US machine for a long time. At the same time, relevant abnormalities can be concealed and a low persistence must be selected, especially for fast-moving structures such as the heart [24].

Contrast resolution

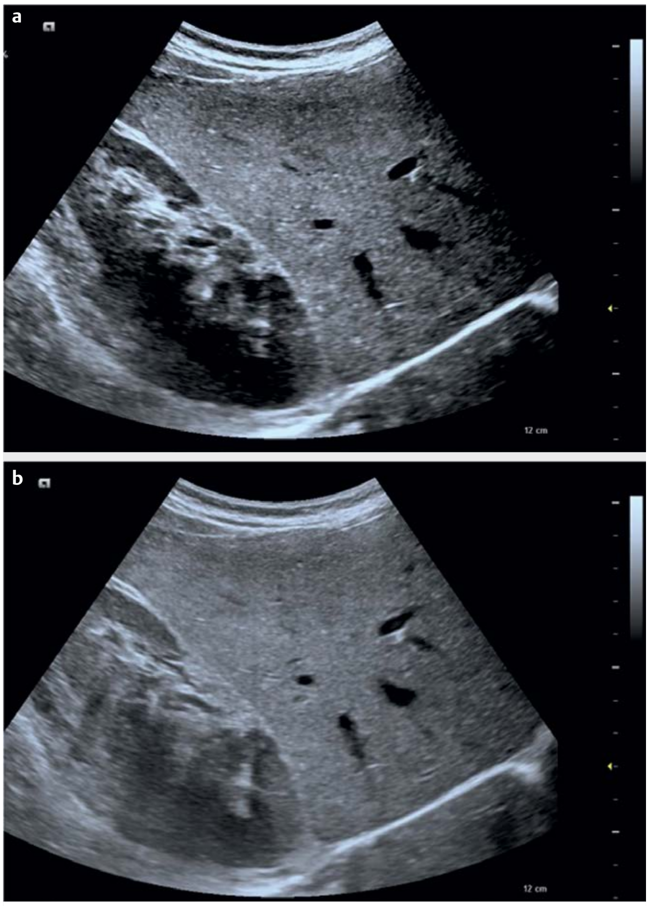
Numerous settings influence contrast resolution and should be adjusted accordingly for optimum image quality. These include nominal frequency, brightness, dynamic range, gray maps/curves and B-color [24].

Dynamic range

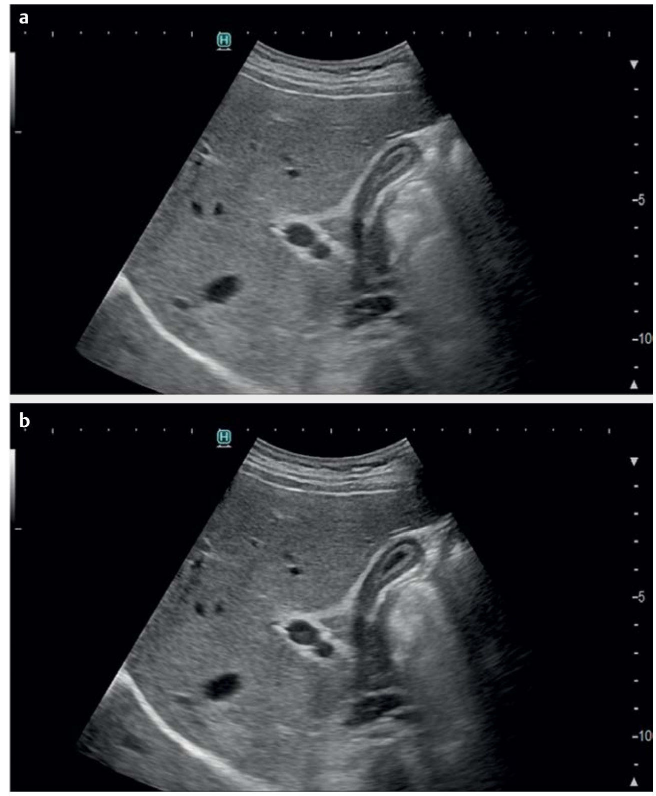
Dynamic range defines the echo strengths shown on the monitor, comparable to the windowing technique in computed tomography. Each received US wave is assigned a gray value and usually displayed on the monitor with 256 gradations. A higher dynamic range would be technically conceivable, especially by the transducer (150 dB), but too broad to be presented on the display. Furthermore, it would be redundant, because the human eye can only distinguish 50–60 of these gradations. A high dynamic range offers more information about the echo patterns, appears brighter and softer and is therefore preferable for representing organ parenchyma. For anechoic imaging of vessels, a low dynamic range is favored. Accordingly, a low dynamic range image results in a more "black-and-white" like shape and thus, higher contrast (▶ **Fig. 8**). Dynamic range is available on live or frozen images.

Gray maps/curves

Whereas dynamic range defines the total number of gray scales displayed, gray maps determine which ultrasonic signal intensity is displayed in which gray scale (how bright/dark). Typically, an S-shaped curve is used instead of a linear correlation. This increas-



▶ **Fig. 8** Imaging of the right liver lobe and right kidney using a low dynamic range with high contrast and coarse echo pattern **a** and high dynamic range with adequate representation of the whole range of parenchymal echos of both parenchymal organs **b**.



▶ **Fig. 9** Imaging of right liver lobe and contracted gallbladder using a linear correlated gray map **a** and an S-shape gray map **b**.

es contrast at intensities that often occur in US images. In general, it is possible to choose between many different gray maps but this is only necessary in rare cases. Gray maps may be adjusted on live or frozen images (▶ **Fig. 9**).

Brightness

The overall brightness of the image can be adjusted by alternating gain or depth dependently via a manual control. Time gain control (TGC)/depth gain control (DGC) compensates for the naturally weaker amplitudes from deeper layers of the image. In standard settings, the depth compensation works well, but must be reduced for structures that hardly attenuate the sound (e. g. liquids). If a high-end machine is used, the adjustments are set automatically where necessary.

The signal gain can also be configured. It controls the image brightness and should be selected so that structures of low echogenicity (e. g. liquids) are displayed in black, and highly echogenic structures such as bones are in white. If this setting is unbalanced, it will lead to a loss of detail due to unused gray scales. When working with an exaggerated signal amplification, the sound lobe diameter analyzed by the device increases but the spatial resolution in the layer thickness plane decreases.

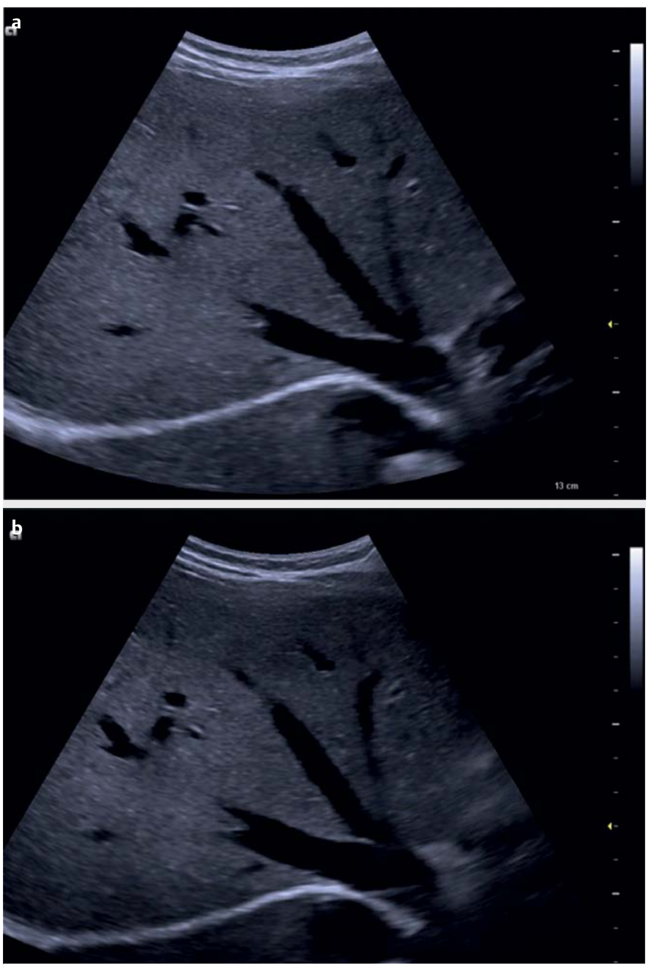
Speckle reduction

Speckle reduction (SR) uses algorithms to reduce the graininess of the image (so-called “speckle noise”). This phenomenon is caused by alternating positive and negative pressure phases in the course of the ultrasonic wave, which causes brighter pixels through superposition and darker pixels through artificial deletion. Smoothing algorithms are used to reduce this undesirable graininess. However, this also means that edges are smoothed and appear less sharp [25, 26]. The SR allows for a more realistic representation with better discrimination of structures. However, details of 1–2 mm in size might be lost.

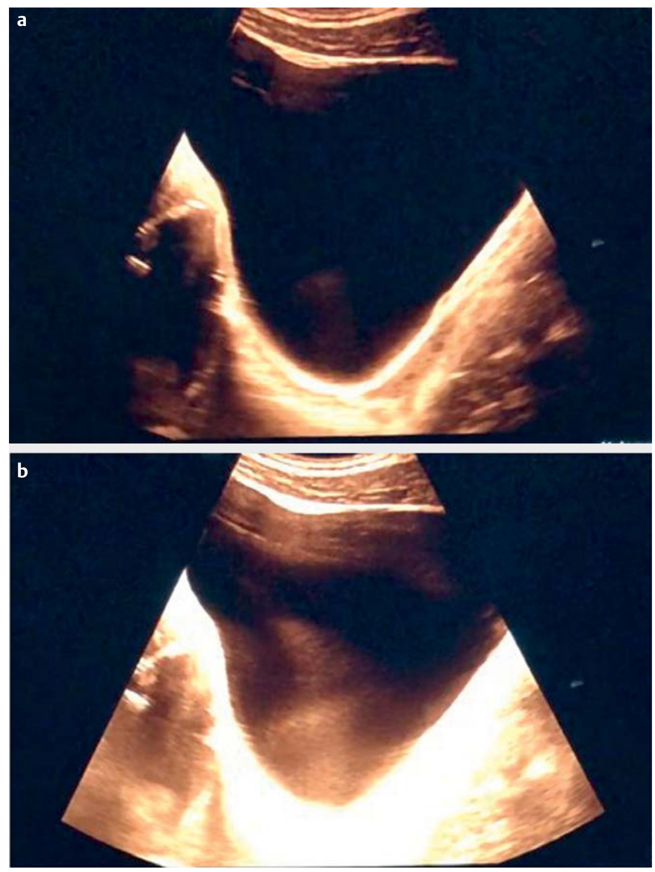
Chromatic colors

The human eye has a significantly higher resolution in color vision compared to black and white. Therefore, it makes sense to display sonographic images in color gradations (monochromatic) or gray scales in different colors (polychromatic). This is highly dependent on the examiner's adaptation to a color scheme. Unbiased students prefer the monochromatic display, opposed to experienced ultrasound users, who are used to black and white images and thus favor this setting [27]. However, the advantage of B-mode colorization over gray mode imaging for the detection of focal liver lesions could not be proven (▶ **Fig. 10**) [28].

▶ **Fig. 10** Different gradations (polychromatic, monochromatic) are shown in grayscales **a** and in different colors **b** and **c**.



▶ **Fig. 11** Examination of the right liver lobe using different settings for transmission power: Energy at 100 % (**10a**) and at 10 % (**10b**). Note the reduction of brightness and loss of detail.



▶ **Fig. 12** Inadvertent effects of automatic image optimization: Manually adjusted image of the filled urinary bladder before **a** and after **b** automatic image optimization.

Other factors

Transmission power

Transmission power describes the energy per unit of time (mW/cm^2) and influences image quality (▶ **Fig. 11**). Depending on transmission power, US applications exert both mechanical and thermal effects on the tissue. Although no adverse effects of diagnostic US have been noticed in several studies, US examination should still follow the ALARA principle ("as low as reasonably achievable"). This is particularly important in fetal US (developing tissue and bone) and in ophthalmology, which is why special examination presets exist for such sensitive tissues [29–31].

Presets

Particular combination settings of the above-mentioned parameters of B-mode imaging can be used in so-called presets, which are either specified by the device manufacturer or can be defined by the user. It is strongly recommended that beginners first develop a feeling and understanding for the corresponding effects of the settings, so that presets can be used in a targeted and situation-specific way.

Automatic image optimization

Most of the newer ultrasonic devices offer the function of automatic image optimization. This can work well in many cases, but also poorly as can be seen in ▶**Fig. 12**. It is usually of benefit to manually adjust the image optimization settings as described in this article and summarized in six steps in ▶**Table 1**. Otherwise, an optimal image setting is not ensured. If the function is nevertheless used, it should be noted that the automatic image optimization function works best if the transducer is held still and the patient does not move.

Trackball or Touchpad

The trackball or touchpad is the mouse of the US device and is the common operating instrument of the screen cursor. The ball can be rotated freely in all axis directions and is used for uncomplicated movements of the cursor on the X and the Y axis on the monitor. Other functions are also controlled with it, such as scrolling through a video, positioning the body marker, or adjusting a scale. The same concept is utilized with a touchpad instead of a trackball.

▶**Table 1** Six steps to achieve optimal settings for B-Mode US.

Steps	B-Mode parameter	Remarks
1st	Transmission power	Depth penetration is improved and scattering is reduced with increasing transmission power; in accordance with the ALARA principle.
2nd	Gain	Signal amplitude is increased and noise is reduced with increasing gain. Adjust as low as possible to avoid overexposure. Use time gain control (TGC)/depth gain control (DGC) if necessary for compensation of strongly enhanced or diminished tissue attenuation.
3rd	Frequency	Spatial resolution is improved at the cost of depth penetration by increasing center frequency (and inversely).
4th	Depth penetration	Adjust to the structure of interest, no higher than required.
5th	Focal zone(s)	At the level of interest or use focus.
6th	Further settings	Only in the case of insufficient image quality: change the preset, adjust the dynamic range, gray maps, persistence, and/or frame rate.

Freeze

This function is used to pause the moving live image to be able to judge individual frames more precisely, or to save and store them. Today's devices also offer the possibility of rewinding a certain period of time. This is particularly advantageous for locating structures that are only briefly visible in the moving image and often elude targeted freezing attempts.

Loop function

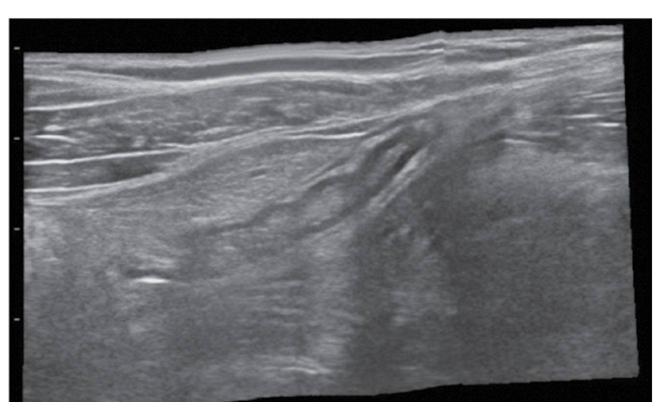
Current devices offer the possibility of rewinding a certain period of video time by continuously storing the captured images, similar to the freeze function above. The length of the loop depends on the system used, while the frame rate is also subject to the converter and image depth.

Panoramic imaging

Due to the limited extension of the acoustic window, depicting large structures and their topographical relations is difficult with traditional US imaging. The examiner usually makes a subjective assessment of the extension of large organs or pathologies. Overview images are not possible with classical US examinations. To overcome this disadvantage of US, panoramic images up to 60 cm long can be produced by merging several images during an even and continuous transducer movement along the structure of particular interest (▶**Fig. 13**). In addition to displaying large organs such as the liver, it can also be used to clearly display the topographical relation and dimensions of tumors, as well as vascular and intestinal tract anatomy. Therefore, panoramic imaging and storage of image loops are very helpful tools to communicate US findings [32, 33].

3D Ultrasound

A clear representation of anatomy and topography can also be accomplished using three-dimensional (3D) US. While the procedure has already been established for prenatal US, precise volumetry,



▶**Fig. 13** The normal appendix depicted using panoramic imaging.

determination of spatial requirements, and topography studies can also improve diagnostic accuracy in abdominal sonography or echocardiography. Meanwhile, reconstruction in real time is also available in the latest devices (4D sonography) [34, 35].

Documentation

Clips

The clips function makes it possible to record short films, and thus display moving structures. Usually motion image data are captured prospectively, while some machines provide the possibility to store films retrospectively. This is feasible because the devices continuously store the live image in a memory and overwrite the oldest images with the newest ones after a certain period of time ("first in, first out"). The length and the supported file formats (e. g. AVI, JPEG and DICOM) depend on the system used, while the frame rate also hinges on the converter and the image depth. It is also possible to save individual images from the clips.

Measurements

The measurement function is an essential part of the US examination, as many diagnostic criteria are based on quantitative findings.

This function is available during an examination or with stored images. The desired measuring points are searched with the trackball or touchpad and set by the "enter" or "set" button. The measurement function includes not only the distance between two points, but also areas, volumes, angles, circumference as well as methods with more complex calculations (gray scale and strain histograms, strain ratio, shear wave velocity, quantification of US beam attenuation and tissue heterogeneity) depending on the examination mode.

Pictogram, body marker

The pictogram is a simple way of making the location of the sound window comprehensible to viewers of the recorded images. Depending on the device, the representation of the pictogram is more or less elaborate and is usually positioned using the trackball/touchpad. In Germany, the use of a pictogram is mandated by health care providers, but a relatively large number of examiners renounce this function [36].

Store

The "store" button allows for permanent storage of single frames of the live image. Different storage media are available ranging from the local hard disk and USB sticks to network storage, where DICOM is the standard format. DICOM stands for "Digital Imaging and Communication in Medicine" and enables communication and data exchange between different systems, such as the ultrasound machine and a viewing workstation or the patient management system. Once transmitted, the content is available on servers or local storage [37]. The DICOM format not only contains the image data but also additional image-related information like patient data, equipment, examination details, and basic image metadata. Nowadays, video clips or cine loops can also be stored, which unlocks additional potential in diagnostics but involves large amounts of data. These large datasets, particularly videoclips which can involve several thousands of images may take up a significant amount of space on Picture Archive and Communications Systems (PACS), which comes at a cost. However, scanners can be configured to only transfer relevant images/video clips selected by the operator.

Print out

This method of preserving images is often used as a quick way of passing them on to patients, but in the course of digitalization it is no longer appropriate for the documentation of examination findings.

Further Literature and Illustrations

We explicitly refer to the further literature on B-mode ultrasound, imaging examples [38, 39], and the guidelines of the European Federation of Societies for Ultrasound in Medicine and Biology (www.EFSUMB.org).

Conflict of Interest

The authors declare that they have no conflict of interest

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