Ultrasound Physics

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KEYWORDS

• Point-of-care ultrasound • POCUS • Ultrasound physics • Image resolution
• Doppler • B-Mode • Artifacts • Knobology

KEY POINTS

- The principles of ultrasound physics involve understanding the properties of ultrasound waves, including their mechanical nature.
- The frequency of an ultrasound wave determines its resolution and the depth of penetration. Higher frequency probes are ideal for imaging superficial structures, and lower frequencies for deeper tissues.
- Propagation speed is dependent on the characteristics of the medium. In general, higher stiffness in the medium results in increased propagation speed, while increased density tends to slow down the ultrasound waves.
- Attenuation artifacts include shadowing, edge artifact, posterior acoustic enhancement, and anisotropy. Propagation artifacts include reverberation and mirror image artifacts.

INTRODUCTION

Ultrasound was first introduced as a diagnostic tool in the mid-twentieth century.[1] Over the past decades, this technology evolved and has become more compact with improved diagnostic quality and advanced features, such as artificial intelligence.[2,3] As the use of ultrasound is becoming more popular, it is essential to have a thorough understanding of the basics of ultrasound physics such as sound wave generation, interaction with tissues, and machine data processing and displays. These concepts have clinical implications at the bedside, especially by differentiating artifacts, minimizing false interpretations, and correctly identifying different pathologies. This article will explore the principles of ultrasound physics, including sound wave dynamics, interactions with biological tissues, and the factors impacting data processing and ultrasound image generation.

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DISCUSSION

Basic Principles

Properties of an ultrasound wave

Ultrasound is a mechanical wave that moves via alternating high-pressure (compression) and low-pressure (rarefaction) phases through a medium (**Fig. 1**).[4,5] Sound particles consist of both transverse and longitudinal waveforms moving within a medium like air, water, or solids. Transverse waves, displace the medium perpendicular to the direction of the wave propagation, while longitudinal waves, displace the medium in a parallel direction. Only longitudinal waves effectively traverse distances, emphasizing their importance in diagnostic ultrasound.[5]

Frequency (Hz): It refers to the number of cycles emitted by the ultrasound probe over 1 second.[5] One cycle of a wave is a complete positive and negative pressure phase. Frequency is determined by the emitting source and is independent of the tissue interacting with the waves. Each ultrasound probe has a characteristic frequency, typically ranging from 1 to 20 MHz. These frequencies far exceed the audible sound range of humans (20 Hz to 20 kHz).[4] Frequencies correlate directly with resolution and inversely with depth (**Fig. 2**).

Period (msec): It is the duration required for a complete wave cycle to occur.[4] Period and frequency are reciprocals of each other.[5]

Wavelength (mm): It is the distance between 2 identical points on the wave (**Fig. 3**). Shorter wavelengths lead to enhanced resolution but reduced depth of penetration.[6] Conversely, longer wavelengths result in a decreased resolution but greater depth of penetration. Wavelength and frequency are also inversely related; the product of this interaction impacts the speed of sound in a medium.[6]

Amplitude (dB): It is the difference between the average and maximum values of an acoustic variable (see **Fig. 3**). It describes the strength of the sound wave. Increased amplitude results in increased echogenicity, or brightness of the signals reflected from an anatomic structure.[7] Amplitude corresponds to the intensity of the signal, indicating its loudness or strength, but it does not affect the speed of sound in medium.

Power (Watts): It is defined as the amount of energy (joules) generated per unit of time, representing the rate at which work is performed.[5] The gain function on an ultrasound machine controls both amplitude and power.

Intensity: It is defined as power delivered over a specific area (Watts/cm^2 or milliWatts/cm^2). Intensity has a direct impact on the thermal bioeffects of ultrasound on human tissue.[5,7] The intensity of the beam can be measured over time (temporal variation) or measured over space (spatial variation). The spatial peak is the point within the ultrasound beam with greatest intensity (highest power divided by smallest area). This is the focal point where the beam is the most focused and intense. Typically, an ultrasound user can adjust the focal point, making the beam stronger or weaker at different points.

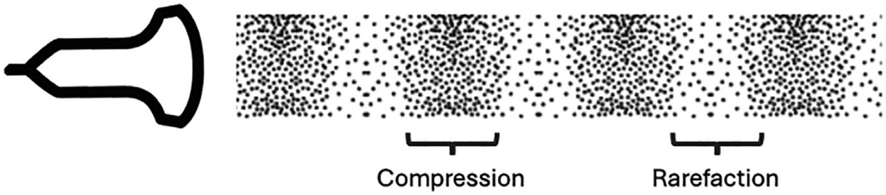

Fig. 1. The phases of ultrasound wave propagation: high-pressure compression and low-pressure rarefaction.

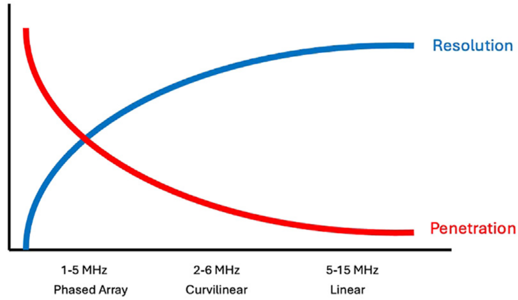

Fig. 2. Relationship between resolution and penetration at various frequencies.

Propagation speed: It is the rate at which ultrasound waves pass through a medium and is independent of frequency. The speed of an ultrasound wave is 1540 m/s in soft tissue. Propagation speed is dependent on the characteristics of the medium. In general, higher stiffness in the medium results in increased propagation speed, while increased density tends to slow down the propagation speed of ultrasound waves.[4–6] For example, though bone is dense, it is also stiff, enabling ultrasound to propagate at its fastest speed of 3500 m/s. In contrast, lung tissue, though dense, is less stiff, resulting in the slow speed of 500 m/s. In general, sound travels fastest in solids, slower in liquids, and slowest in gases.

Principle of Ultrasound Probes

Piezoelectric crystal probes

The active elements of ultrasound probes are the piezoelectric crystals which have the capability to convert electrical current into mechanical pressure waves or ultrasound waves.[5] These crystals vibrate when an electrical voltage is applied across them, subsequently generating sound waves. Piezoelectric crystals can convert returning echoes back into electrical energy then into images.[5] The piezoelectric crystal switches between transmitting and receiving ultrasound waves and spends 99% of

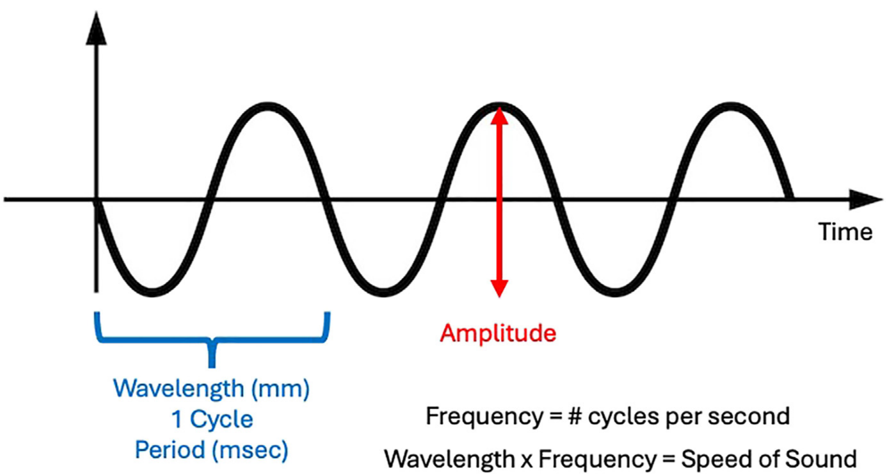

Fig. 3. Properties of a sound wave.

the time receiving echoes, which are then converted into images.[8–10] This phenomenon is described by the pulse-echo-principle, in which ultrasound uses short pulses of sound to create images.

The ultrasound wave generated by the probe is referred to as the beam. The beam is cylindrical in the near field (ie, close to the probe) and transitions to a conical shape in far field. A higher resolution requires a narrow cross-sectional area of the beam and an elongated near field.[6] To elongate the near field, the frequency must be increased, and wavelength decreased, or the size of the probe needs to be increased. The beam can be focused to create a narrower beam at a chosen depth, resulting in enhanced resolution (**Fig. 4**). The different arrangements of crystal combined with the unique footprint of each probe manipulate the ultrasound beam to create diverse images.

The bandwidth of a particular probe is the range of frequencies at which the probe will operate. Probes emit pulses of sound waves with a set duration (ie, pulse duration) which is the time from the start to the end of a pulse, also called transmit time.[5] In the interval between pulses, the probe listens for the return of these reflected pulses, called the “receiving time.” Shorter pulses enable discrimination of smaller objects, resulting in higher resolution. The spatial pulse length is the length of each pulse.

The pulsed repetition period (PRP) is the time between the start of 2 pulses and includes 1 transmit time plus 1 receiving time. Pulsed repetition frequency (PRF) is the number of pulses generated in a single second. PRF and PRP are inversely related (**Fig. 5**).[4] A higher PRP results in an extended “listening” time from the probe which allows for deeper imaging. Deeper structures need sound waves to travel further thus taking longer to return, requiring longer listening times. A higher PRF means more pulses in a second, more cycles thus shorter listening times, producing better image resolution at shallow depths.

Silicon chip probes

In some handheld ultrasounds, a silicon chip replaces the traditional piezoelectric crystals. This novel approach uses the "capacitive micromachined ultrasonic transducer" (CMUT) consisting of a thin membrane suspended over a cavity or gap.[11,12] Together, the membrane and the base underneath form what is called a capacitor.

When a voltage is applied to the system, electrostatic forces pull the membrane toward the base, narrowing the gap between them. This quick movement creates ultrasound waves.[12] By adjusting the voltage, the device can produce ultrasound waves at various frequencies. Just like with traditional crystal-based probes, the echoes that bounce back cause the CMUT membrane to move. This movement changes the capacitance, which is then converted into an electrical signal and subsequently an

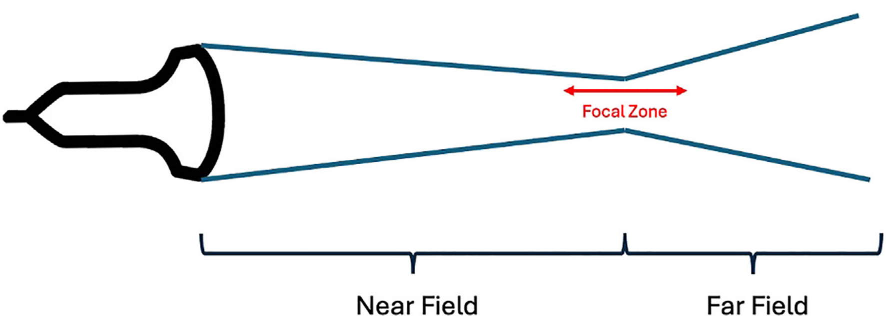

Fig. 4. Description of an ultrasound beam.

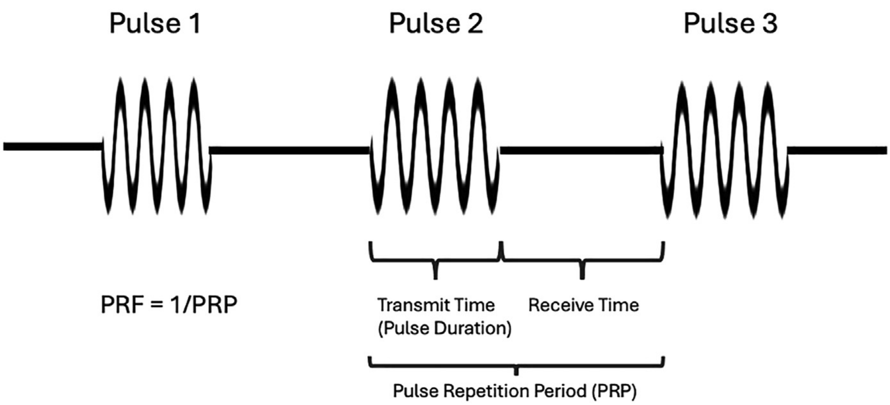

Fig. 5. Pulsed repetition frequency, pulsed repetition period, pulse duration/transmit time, and receive time.

image. This chip can individually control each CMUT element, enabling different imaging modes without requiring multiple probes.[12]

Ultrasound Interaction with Tissues

As the ultrasound beam passes through different tissue, it undergoes processes that result in the reduction and weakening of its energy.[7] This is known as *attenuation.*

Attenuation refers to the gradual decrease in intensity and amplitude of a sound wave as it propagates through a medium.[5] Sound weakens more as it travels longer distances, such as through deeper tissue, and when emitted at higher frequencies. Thus, deeper imaging and higher frequencies attenuate the most whereas shallow imaging and lower frequencies exhibit less attenuation. Therefore, the highest resolution images are typically achieved by focusing on the shallower depths (Fig. 6).

Each type of tissue has its own effect on attenuation which is quantified by their attenuation coefficient (dB/cm). The attenuation coefficient of soft tissue is about half of the probe frequency (about 0.5 dB/cm/MHz).[5] Water demonstrates minimal attenuation at frequencies of 10 MHz or less, making it an excellent acoustic window.[5] When ultrasound sound waves are emitted from the probe, they travel into the body and interact with various tissues. Two main things can happen to these sound waves: 1- Reflection back to the probe, 2- Attenuation via absorption, refraction, scatter, or reflection away from the probe.

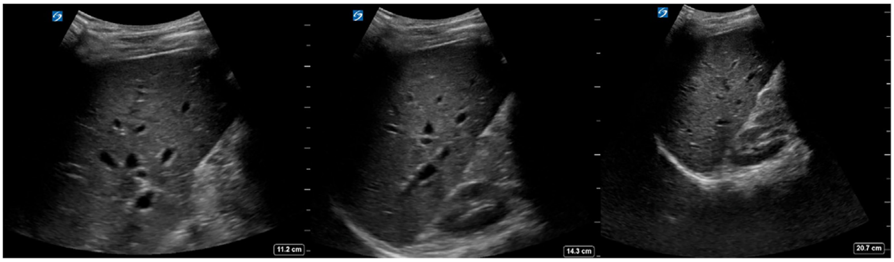

Fig. 6. Appearance of the liver at different depths, with highest resolution images at shallower depths.

Reflection

Reflection describes the process of waves bouncing back toward the probe source. The angle of incidence is the angle at which ultrasound waves engage with any structure, and is identical to the angle of reflection. A perpendicular angle of incidence is ideal, so reflected waves return to the probe in greatest concentration (**Fig. 7**).[4] Not all reflections help create an image. Some sound waves reflect away from the probe in different directions, which means they are attenuated or lost for creating images.

There are 3 primary factors influencing reflection as follows:

i. Tissue Density: Different tissue densities result in different degrees of reflection. Tissues with higher density reflect more sound, resulting in hyperechoic or brighter images (eg, bone, dense foreign bodies).[4,6] A low-density medium, such as fluid, exhibits poor reflection resulting in anechoic ultrasound images.
ii. Interface Type: Smooth interfaces return a higher proportion of waves to the probe. In contrast, irregular interfaces cause sound waves to scatter away from the probe, with only a portion returning to the probe, ultimately reducing the image quality.[4]
iii. Acoustic Impedance: Acoustic impedance is an intrinsic property of each type of tissue which affects the strength of the reflected sound wave. It is the product of tissue density and speed of sound in the medium.[7] The difference in acoustic impedance between the interfaces of 2 medium determines the amount of energy that travels through the second medium (transmission), as well as the amount of reflected energy. At perpendicular incidence, the difference between the impedance of the 2 media is the only factor that affects reflection. If the difference in impedance between the 2 media is large, sound waves are completely reflected (**Fig. 8**). For example, since air has a much lower impedance than either the probe or skin tissue, sound waves that contact air are almost completely reflected, resulting in minimal transmission of sound beams.[5] By providing a medium between the probe and the skin in the form of ultrasound gel, more sound waves are allowed to pass from the probe into the body.

Absorption

When ultrasound waves are scattered and their energy is absorbed by tissues, it generates vibration energy and heat. Absorption increases with increased probe frequency and greater depths (**Fig. 9**). Under normal diagnostic scanning conditions,

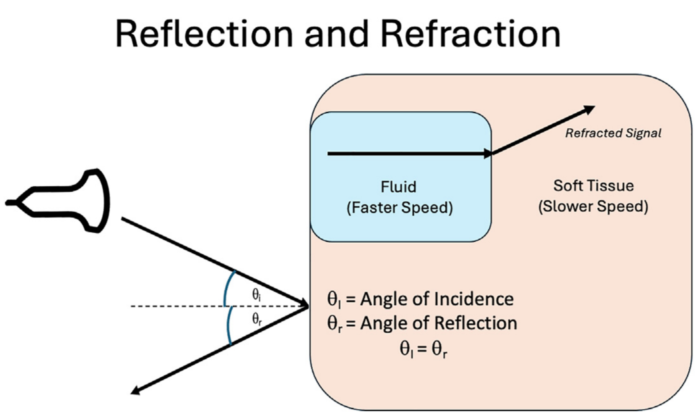

Fig. 7. Reflection: angle of incidence = angle of reflection. Refraction: wave bends from its original straight path due to difference in propagation speeds.

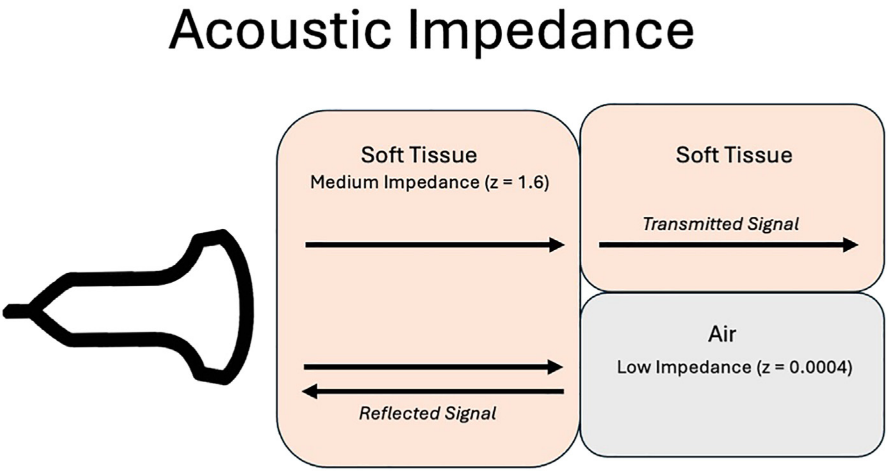

Fig. 8. Reflection occurs when the difference in impedance between the 2 media is large.

the amount of heat produced is too small to cause any significant temperature changes.[5]

Refraction

Refraction refers to the bending of a wave from its original straight path due to 2 media having different "stiffness," resulting in a change in propagation speeds (see Fig. 7).[8,9] The original angle of incidence and the difference in the propagation speed of the 2 media determine the ultimate angle of the refraction. The amount of deflection is proportional to the difference in stiffness between the 2 media.[5]

Scatter

When sound waves encounter a medium with a heterogenous surface, most of the original wave continues to travel along its original path but a small portion is scattered in random directions.[5] Sound scatters when the tissue interface is smaller than the wavelength of the sound wave it receives. For example, lung tissue scatters sound waves due to air-filled alveoli. The degree of scatter is also directly proportional to frequency; high- frequency sound scatters more than low-frequency sound. In point-of-

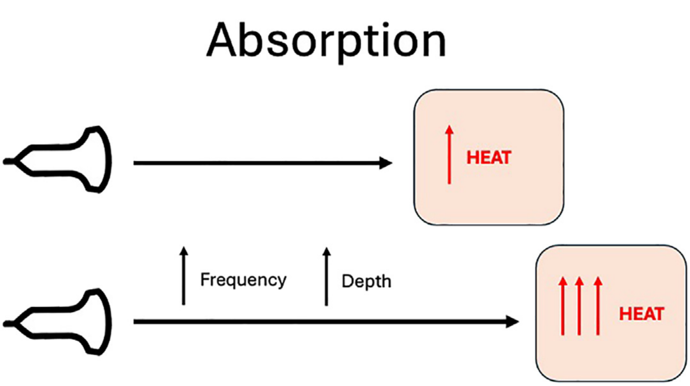

Fig. 9. Absorption increases with increased frequency and depth.

care ultrasound (POCUS), the goal is to use the highest frequency possible to produce the clearest images while limiting the increase in scatter.

Image Resolution

Resolution refers to how well sound waves can differentiate 2 closely spaced objects. In diagnostic ultrasound, 2 types of resolution are described: *spatial and temporal.*

Spatial resolution

Divided into axial (parallel to the beam) or lateral (perpendicular to the beam) resolution (**Fig. 10**).[5]

i. Axial resolution distinguishes structures along the main axis of the beam. Higher frequencies, shorter wavelengths, shorter pulses, and smaller spatial pulse lengths all result in higher resolution.
ii. Lateral resolution refers to the minimum distance that 2 structures are separated when they are perpendicular to the beam. Lateral resolution is highest at the narrowest portion of the beam and decreases as distance from focal point increases.

Temporal resolution

Temporal resolution is the time it takes to create an image and is measured by the number of frames created per second (ie, frame rate). Detailed objects slow the frame rate due to a longer time required to make each frame. Shallow images require less time to create since the beams traverse a shorter distance, resulting in faster frame rate and improved temporal resolution.[5] With increased depth, the ultrasound probe spends more time listening to collect data before displaying the image, resulting in a reduced frame rate and lower temporal resolution.[7] The size of the probe matters as well. The smaller footprint of the phase-array probe saves computing power, resulting in a higher temporal resolution which is critical for imaging an actively beating heart. Stationary objects, such as the liver, require less temporal resolution and can be imaged with a larger, more detailed curvilinear probe.

Types of Probes

Linear array probes (**Fig. 11**)*:* They have a sequence of piezoelectric crystals arranged in a line, wired to all work simultaneously, along a flat footprint.[6] Linear probes are generally used clinically for superficial structures and are designed to produce higher frequencies and higher resolution.

Curvilinear probes (see **Fig. 11**): They contain crystals arranged in a curved fashion with a large footprint.[7] The larger footprint typically produces increased lateral resolution. A variety of convex array probes exist.[7] Endocavitary probes have a particularly wide field of view, up to 180°, and are placed close to the organ of interest with high frequency, producing excellent tissue resolution. Though most commonly used for

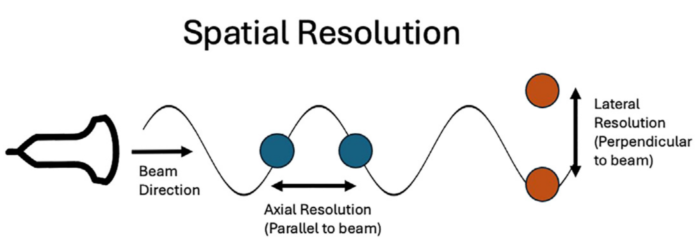

Fig. 10. Spatial resolution refers to either axial or lateral resolution.

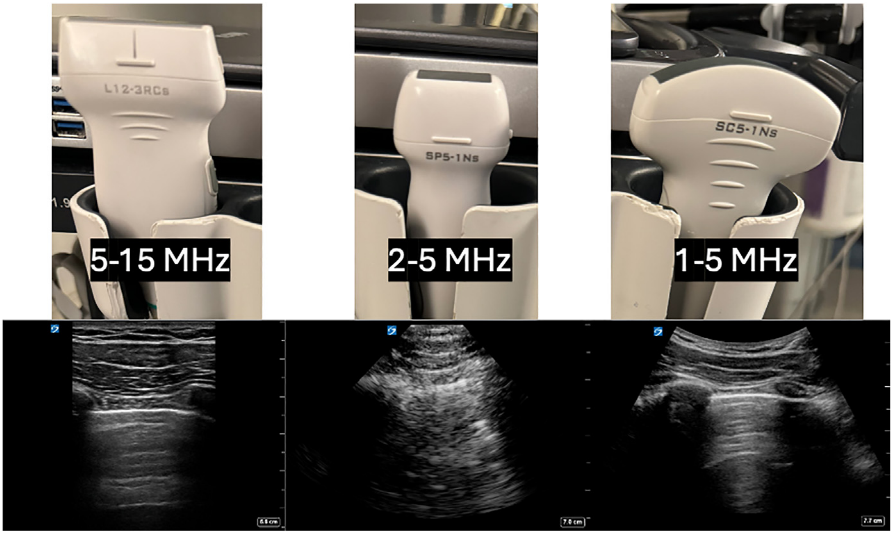

Fig. 11. Linear, phased-array, and curvilinear probes with their corresponding frequencies and appearance in lung ultrasound (*from left to right*).

pelvic scanning, similar endocavitary probes can be used for intraoral examinations such as the diagnosis of peritonsillar abscess.[7,13]

Phased array or sector probes (see Fig. 11): They contain crystals grouped into a tiny cluster on a small, flat footprint which produce a pie-shaped sector image.[6] The small footprint allows imaging in between ribs, making it ideal for cardiac examinations. Phased array probes tend to have lower frequencies than linear probes but high temporal resolution and Doppler capabilities, making them ideal for cardiac POCUS.

Modes of ultrasound

Ultrasound data are displayed through 3 main modes: B-mode, M-mode, and Doppler mode.

Brightness Mode

Brightness mode (B-mode) translates the multiple amplitudes of waves into dots of varying intensity which generate anatomic images on the ultrasound screen.[5] In B-mode, increased depth results in increased time of travel of the pulse. Weaker reflections yield darker dots while stronger reflections produce brighter dots. An anechoic image represents an absence of returning sound waves (no reflection). Darker (hypoechoic) images represent few returning sound waves (weaker reflection). Bright (hyperechoic) images are produced by large amplitude of returning sound waves (stronger reflection).

Motion Mode

Motion mode (M-mode) displays the movement of structures by capturing returning sound waves in 1 line of the B-mode image displayed over time (Fig. 12). All motion that occurs along that line is captured and displayed on the screen. Stationary objects are depicted as straight lines while objects in motion are represented by sinusoidal or

Fig. 12. Pleural lining of the lung in brightness mode (*left*) versus motion mode (*right*).

grainy lines.[5] M-mode is highly useful for clinical scenarios requiring a graphical representation of fine movement (eg, pleural sliding, measuring fetal heart rate).

Doppler Mode

Doppler analyzes and displays the direction, pattern, and velocity of blood flow and direction or speed of tissue motion. The Doppler shift (or frequency shift) is the change in frequency between the sent and the returning sound waves.[6] The signal processing of the Doppler shift is called spectral analysis.[6] A sound source traveling toward a listener appears to have a higher pitch while a sound source moving away from a listener appears to have a lower pitch. This difference or shift in frequency is the Doppler shift. Clinically, Doppler shifts are created when actively moving red blood cells are struck by transmitted sound waves, generating scatter which is then analyzed (ie, Rayleigh scatter).[5,6] The change of frequency of shift correlates with velocity and direction of particle motion. Doppler shift is directly proportional to the velocity of blood and cos θ (**Fig. 13**). A lower frequency is preferred with Doppler to allow measurement of higher flow velocities. No Doppler shift can be recorded at a perpendicular incidence to the flow, whereas the greatest Doppler shift occurs when the beam is parallel to the flow.[5] Doppler is divided into 3 types: color Doppler, power Doppler, and spectral Doppler.

Color Doppler

Color Doppler visually depicts direction and velocity of blood flow by color-coding the corresponding B-mode image (usually in blue and red) (**Fig. 14**).[8,9] Notably, red and blue do not represent arteries and veins, rather they represent the direction of blood flow. On the ultrasound screen, a scale is displayed when color Doppler is activated, demonstrating that the color at the top of the scale represents flow toward the probe while the color at the bottom of the scale represents flow away from the probe.[7] The velocity of blood flow is demonstrated by varying intensities of the red and blue colors. The highest measurable velocity beyond which the Doppler will misrepresent the velocity and direction of blood flow is called the Nyquist limit.[6] Aliasing is a phenomenon wherein blood flow is depicted in the opposite direction with the alternate color when the blood flow velocity surpasses the velocity representation of the color key. Color aliasing can be mistaken for reversal of blood flow, or regurgitation, particularly in the clinical situation of mitral stenosis.[6]

Power Doppler

Power Doppler depicts only the amplitude of returning echo frequency shifts and ignores flow velocity or direction.[8,9] This allows for the detection of movement in very low flow states such as in testicular or ovarian torsion (**Fig. 15**).

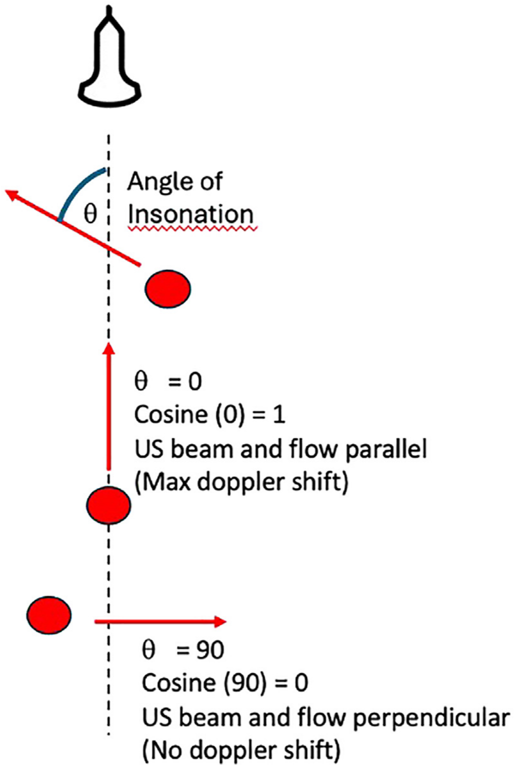

Fig. 13. Doppler shift ~ Velocity of Blood × cos (theta).

Spectral (or quantitative) Doppler

Spectral (or quantitative) Doppler generates a graph with velocity of the moving structure on the vertical axis and time on the horizontal axis.[7] There are 3 common types of spectral Doppler: pulsed-wave, continuous wave, and tissue Doppler.[8,9]

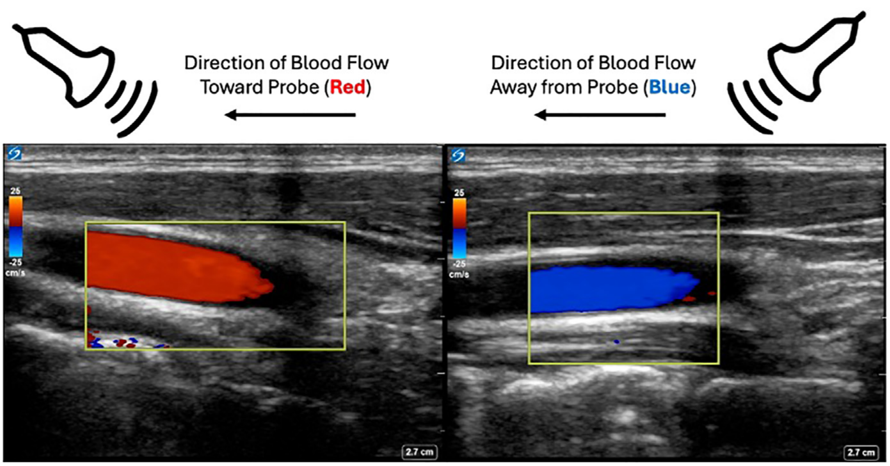

Fig. 14. Changing the angle of the probe in carotid artery ultrasound results in the blood flow appearing either red (toward the probe) or blue (away from the probe).

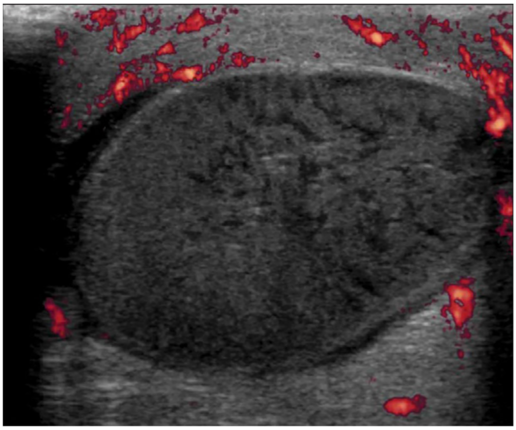

Fig. 15. Power Doppler showing flow in the surrounding tissue but absence of flow in the testicle, compatible with testicular torsion.

i In pulsed-wave Doppler, the probe sends ultrasound pulses to a predetermined depth at the location of the Doppler sample gate on the screen, then listens for returning beams to determine flow velocities at that specific location (intermittent sampling) (Fig. 16). The "spectrum" of returning echoes is plotted. Venous flow appears more continuous, while arterial flow is more triangular with varying amplitudes of systole and diastole. In pulsed-wave Doppler, only 1 crystal is required for both transmitting and receiving signals in between pulses.[5] Pulsed-wave Doppler is commonly used in cardiac examinations (Fig. 17A).

ii In continuous wave Doppler, the probe continuously sends and receives reflected waves (continuous sampling) and can detect high-frequency signals.[6,8,9] One crystal transmits continuously while another crystal receives continuously.[5] Velocities are measured along the entire line of interrogation. These characteristics allow for

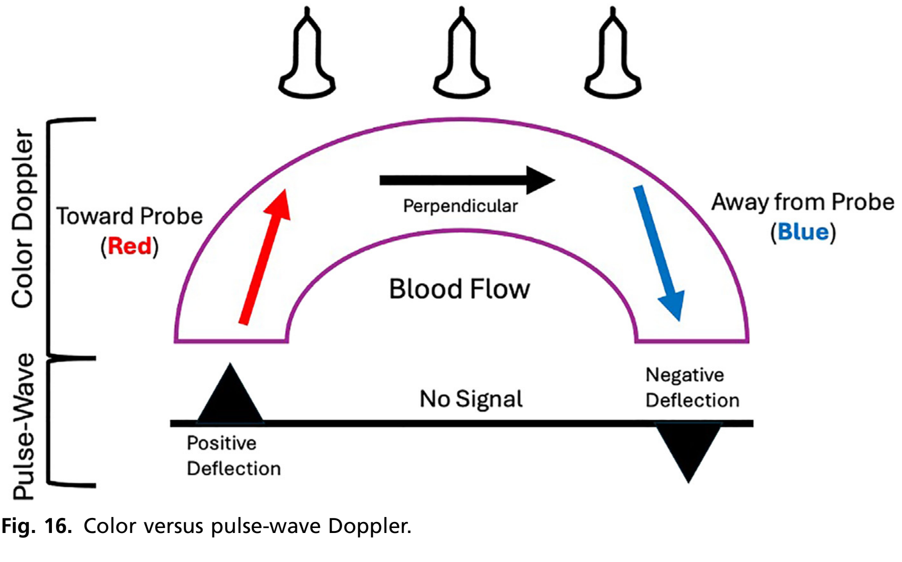

Fig. 16. Color versus pulse-wave Doppler.

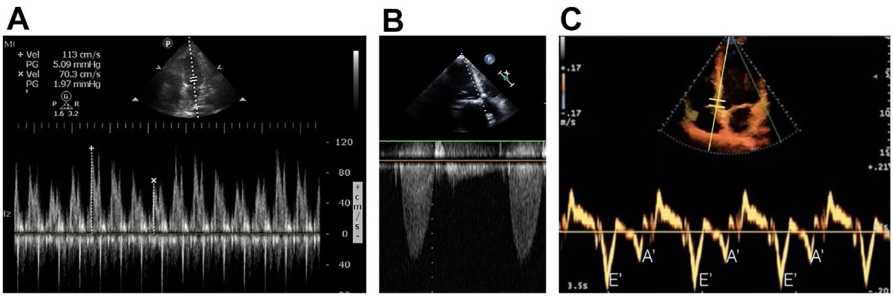

Fig. 17. (*A*) Pulse-wave Doppler analysis of mitral valve inflow velocity. (*B*) Continuous wave Doppler analysis of the aortic valve. (*C*) Tissue Doppler analysis of the mitral valve annulus movement (E′) to evaluate diastolic dysfunction.

the analysis of blood flow at high velocities such as through stenotic cardiac valves, but the drawback is poor depth discrimination (Fig. 17B).

iii Tissue Doppler describes the use of Doppler to interrogate tissue movement.[8,9] A clinical application of tissue Doppler is the measurement of mitral valve annulus movement to evaluate diastolic dysfunction (Fig. 17C).

Image artifacts in clinical scenarios

The physics of ultrasound image generation allows a greater understanding of artifacts. Though artifacts are technically considered an error in imaging, they can be used to obtain key information about anatomy and pathology. The 2 main categories of artifacts are attenuation and propagation artifacts.[14]

Attenuation Artifacts

Shadowing

Shadowing occurs when a much weaker signal returns from behind a strong reflector.[8,9] Structures of increased density are highly reflective, resulting in most of the sound waves returning to the probe and allowing almost no waves penetrating behind the structure, creating a shadowing effect. Common examples of these strong reflectors are bone, metal, plastic, wood, glass, and calcium stones (Fig. 18). Air

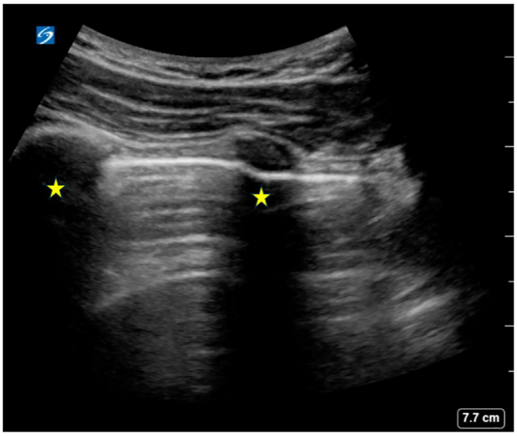

Fig. 18. Rib shadowing (*yellow stars*) in lung ultrasound.

interfaces are also highly reflective but typically create less discernible shadows. Air interfaces with shadowing are usually found in the lungs and bowels, creating “dirty” shadows (Fig. 19).

Edge artifact

Edge artifact describes a thin acoustic shadow behind the lateral edges of cystic curved structures, commonly found clinically on the edges of the gallbladder and bladder. Sound waves striking a curved structure at angles other than 90° result in refraction or bending of the waves off the surface and few echoes returning to the probe from an area expected to reflect echoes, resulting in a shadow.[4] Clinically, edge artifact can appear like pathology. For example, edge artifact can be difficult to differentiate from the shadowing behind a calcified gallstone. However, if following the shadow to its origin reveals a hyperechoic structure, then this more strongly supports a gallstone rather than edge artifact (Fig. 20).[7]

Posterior acoustic enhancement

Posterior acoustic enhancement describes the phenomenon in which sound waves travel with less attenuation through a fluid-filled structure than the surrounding tissue, and return to the probe with a higher amplitude than the adjacent sound waves that pass through soft tissue, resulting in the area behind the fluid-filled structure appearing more hyperechoic relative to the adjacent soft tissue (Fig. 21).[4,7]

Anisotropy

Anisotropy is a specific type of attenuation artifact which refers to the directional dependence of structures on attenuation. Sound attenuation increases when the waves travel perpendicular to the structure and decreases when traveling parallel. This property is most often seen in tendons and results in the tendon occasionally appearing dark and being mistaken for vascular or fluid-filled structures (Fig. 22).

Propagation Artifacts

Reverberation

Reverberation occurs when sound waves encounter 2 highly reflective layers, causing sound to bounce back and forth between the 2 layers before returning to the probe.[4] Some of the sound waves return to the probe as expected while others that are repeatedly reflected take longer to return. The probe detects a prolonged travel time from these waves and assumes the longer travel time is a reflection coming from a structure

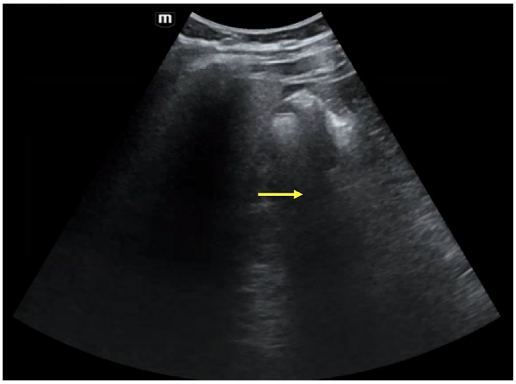

Fig. 19. "Dirty shadowing" (*yellow arrow)* from bowel gas.

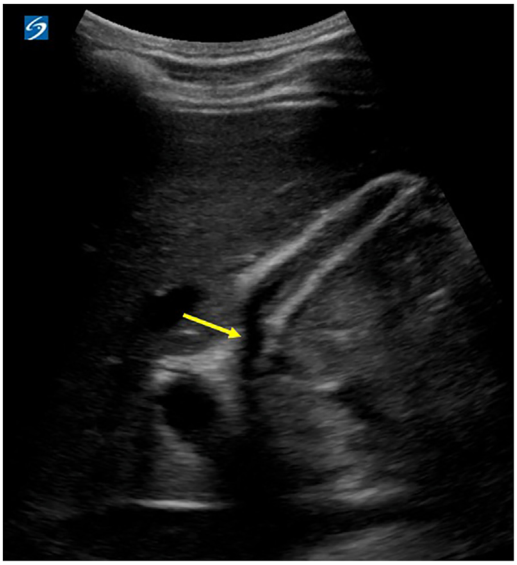

Fig. 20. Edge artifact (*arrow*) in a gallbladder.

deep to the original structure, thus displaying additional deeper "reverberated" images. The first and second reflections closest to the probe are real but the remaining do not correspond to true anatomy.[5] A clinical example of this is A-lines and B-lines in lung ultrasound (Figs. 23 and 24).

Mirroring

Mirroring occurs when a duplicate image of a structure is depicted on the opposite side of a strong reflector, typically deeper than the real structure.[4,5] This artifact commonly occurs at the diaphragm. When waves reflect off the diaphragm and hit another structure like the liver, these waves then return to the strong reflector before finally returning to the probe, resulting in a longer travel time than the waves that take a

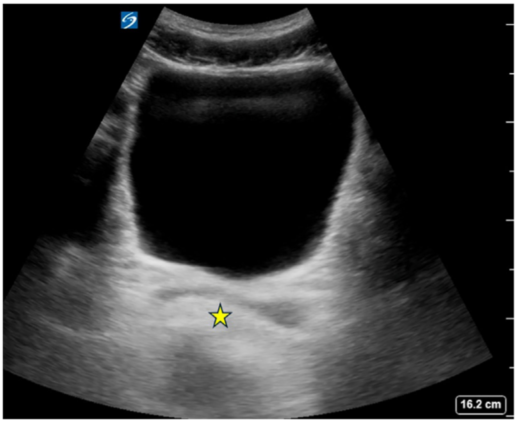

Fig. 21. Posterior acoustic enhancement (*star*) in bladder ultrasound.

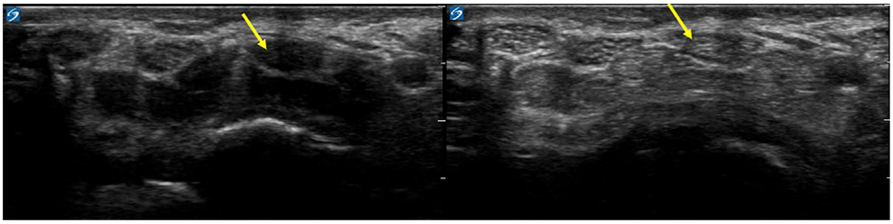

Fig. 22. Anisotropy causing a wrist tendon (*arrow*) to appear anechoic which becomes hyperechoic after changing the angle of the transducer.

single round trip from the probe to the liver and directly back to the probe. The additional travel time is depicted by the probe as an additional structure deep to the strong reflector since additional travel time is depicted as increased depth (Fig. 25).

Side lobe artifact

This artifact is caused by low energy, off-axis "side lobes" of the main ultrasound beam encountering a highly reflective object, then returning to the probe with enough strength to be mistakenly "assigned" to the main central beam and displayed at a false location.[7–9] This artifact usually occurs in anechoic structures and appears as hyperechoic lines inside such as the bladder or gallbladder (Fig. 26). Changing the angle of the probe and rotating the probe to see the structure in multiple views can differentiate artifact versus pathology.

Knobology and Optimizing Image Quality in Clinical Settings

While ultrasound machines can vary in their interface and complexity, they often share similar basic controls (Fig. 27).[4,7–9] Depth and gain are 2 core functions to optimize

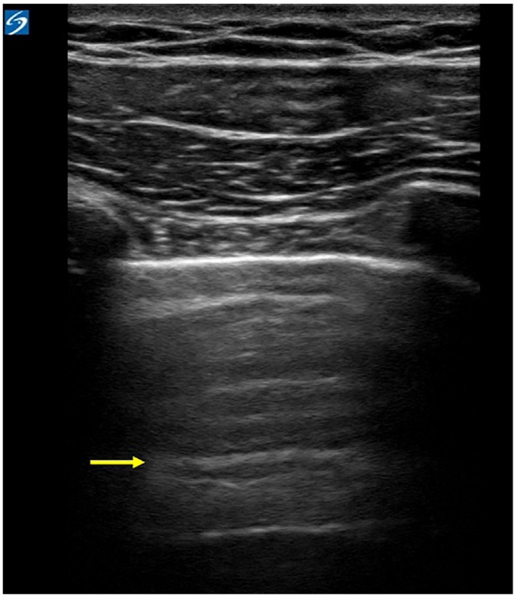

Fig. 23. A-lines (*arrow*).

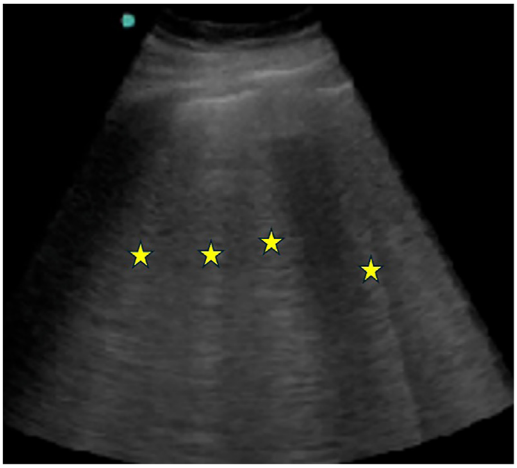

Fig. 24. B-lines (*stars*).

images. Most of the time, one should use only the depth necessary to see the structure of interest, since increasing depth decreases the resolution of the entire image. The right side of the most ultrasound screens will have lines corresponding to the depth in centimeters.

Adjusting gain changes the overall strength or intensity of returning echoes, making the image more hyperechoic or hypoechoic. Time Gain Compensation is an additional ultrasound feature that reduces the impact of tissue attenuation. Time Gain Compensation allows the user to change the strength of returning echoes at various depths, which is particularly important to compensate for posterior acoustic enhancement, which could mask pathology (eg, free fluid posterior to the bladder).

The focal zone of the ultrasound beam can also be adjusted with the “focus” button, allowing image quality to be improved at specific depths. Resolution is best in the focal zone.

Tissue harmonics refers to the tendency of tissue to resonate at multiples of the incident frequency transmitted by the probe.[4] The probe can be adjusted to receive the incident frequency and harmonic frequencies using the tissue harmonic imaging

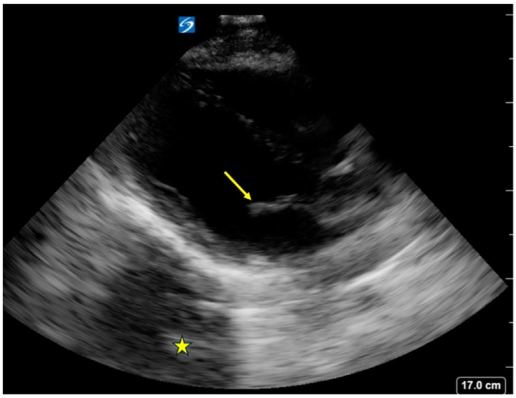

Fig. 25. Mirror image artifact: The pericardium acts as a strong reflector, causing the actual mitral valve (*arrow*) to be seen again as a deeper duplicated artifact (*star*).

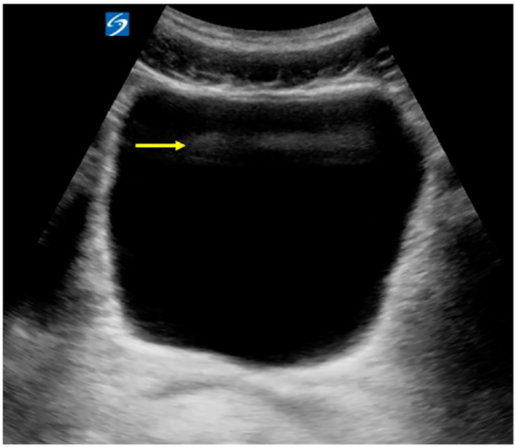

Fig. 26. Side lobe artifact in bladder ultrasound.

function, and then combining the 2 to create an image of higher resolution. Using the tissue harmonic settings results in less scatter and artifacts, producing cleaner images.[7]

Safety

As the intensity or energy of an ultrasound beam increases, the amount of heat directed at a particular anatomic area increases, which could theoretically cause tissue injury. No studies to date have proven that diagnostic ultrasound has damaging effects on tissues, including fetal tissue.[7] However, users of ultrasound still abide by the as low as reasonably achievable (ALARA) principle, which refers to use of the lowest possible power output necessary for proper image development.[7] Clinically, the ALARA principle is particularly important for obstetric and ocular examinations. The obstetrics and ocular presets should always be selected for these examinations to automatically maintain the power output below the levels approved by the United States Food and DrugAdministration for these tissue types.[7]

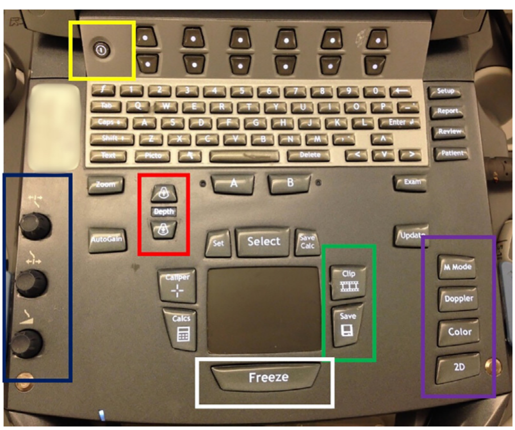

Fig. 27. Sample ultrasound control pane. Power (*yellow*), depth (*red*), gain/time gain compensation (*blue*), various modes (*purple*), save (*green*), freeze (*white*).

SUMMARY

Understanding the physics of ultrasound and the science behind image generation including wave properties, probe technology, and sound interaction with tissue is a critical component of patient care, especially as POCUS is becoming more widely used in diverse settings. With this foundation, clinicians are better equipped to optimize the clinical utility of ultrasound for timely and accurate diagnosis and management.

CLINICS CARE POINTS

- Lower frequencies penetrate deeper into tissue but result in lower resolution images. Higher frequencies penetrate with less depth but result in higher resolution images.
- Attenuation occurs due to absorption, reflection, refraction, and scatter of sound energy away from the probe.
- Common artifacts include shadowing, edge artifact, posterior acoustic enhancement, and anisotropy, reverberation, mirror image artifacts, and side lobe artifact.
- Depth, gain, time gain compensation, focus, and tissue harmonics are all functions that allow the optimization of ultrasound images.
- Users of ultrasound should abide by the ALARA principle, which refers to use of the lowest possible power output necessary for proper image development.

DISCLOSURE

The authors have nothing to disclose

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