Surgery Open Science 18 (2024) 123–128

Contents lists available at ScienceDirect

Surgery Open Science

journal homepage: www.journals.elsevier.com/surgery-open-science

Special Section on HPB Surgical Ultrasound; Edited by Dr. Ellen Hagopian

Ultrasound principles and instrumentation

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ARTICLE INFO

Keywords:
Ultrasound
US
Biliary
Pancreas
Liver
Physics

ABSTRACT

Ultrasound (US) is a fundamental and inexpensive tool both for the prompt diagnosis and for the study of diverse medical conditions. Its widespread use is partly due to the availability of US devices in the daily practice of physicians. US can be performed in real-time and is instrumental in the generation of clinical algorithms for the management of situations like trauma. It also constitutes a primary approach for the study of oncological diseases, and a guidance tool for interventions such as percutaneous drainages. In addition, and specifically for HPB surgeons, US is an essential tool in the operating room: intraoperative (either open or laparoscopic) US is necessary for the accurate determination of the stage, location, number, and margins of tumors within the liver, pancreas, or biliary tree. On another note, reading and understanding US images are skills that require time and training, which should be taught during surgical residencies. However, this is not customary in most residencies globally. This chapter offers a concise yet comprehensive elucidation of the basic principles of ultrasonography, the instruments required to perform an ultrasonic assessment of a patient, and the basic ultrasound controls.

Introduction

Physics

Sound

US is a type of non-ionizing radiation used to acquire real-time images of tissues [1] [2]. It corresponds to the sound energy portion of the spectrum, above the audible range of 20 Hz to 20 kHz, and is transmitted through a medium as a mechanical wave. Sound can be propagated through different materials (like tissues and liquids within the body) by collision with different particles within a vibrating object, without displacing them from their original position. Sound has the same physical characteristics as all waves. Frequency is measured in Hertz (Hz), with 10 Hz being ten complete oscillations in one second. The higher the frequency of the wave, the higher the pitch of the sound. A healthy human ear can hear sound between 20 and 20,000 Hz (20 kHz, kHz). US cannot be heard by humans, since its frequency ranges between 2 and 18 MHz (MHz). US, as a medical diagnostic tool, ranges between 2 and 15 MHz. The wave has a crest or peak, and a trough, which is the lowest part of the wave in terms of energy. The amplitude is the distance from the trough to the crest. The wavelength is the distance between two crests, and is inversely proportional to the frequency, thus, the higher the frequency, the lower the wavelength (Fig. 1). This concept is important when selecting the appropriate instrumentation and settings. A higher frequency results in better image resolution, but with lower depth penetration. For example, a 3.5 MHz probe is appropriate for transabdominal US (TAUS), as it will have deeper tissue penetration (more than 15 cm), but will have lower image resolution. In contrast, in intraoperative US (IOUS) a 7.5 MHz probe is a more appropriate choice where image resolution is of greater importance than depth of penetration (typically less than 8 cm) [3,4].

US image

The US image is generated by the exchange of electrical and sound energy by piezoelectric crystals located in the US probe (piezoelectric effect). When electrical energy is applied to the US probe, piezoelectric crystals convert energy into sound energy, which is emitted from the US probe as sound waves. When sound reaches its target (such as tissue, etc.), it is reflected back (some energy is lost by absorption, scatter, refraction; see below) to the probe. The reflected waves are captured by the probe and the piezoelectric crystals convert the sound energy back into electrical energy. The processor transforms this energy into an ultrasound image. The conversion of electrical to sound energy and back is

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https://doi.org/10.1016/j.sopen.2024.02.005
Received 1 June 2023; Received in revised form 10 February 2024; Accepted 19 February 2024
Available online 11 March 2024

the piezoelectric phenomenon.

Every tissue has different properties that determine the opposition they offer to the propagation of US waves, a characteristic known as impedance which is crucial for the creation of the US image. Liquids or tissues with high water composition have low impedance whereas air and solid tissues (gas and bones) exhibit a higher impedance. The impedance difference between both tissues in an interface will determine how sound is reflected. For example, there is a high impedance difference between muscle and bone, resulting in a high percentage of US reflected to the probe. On the other hand there is a minimum impedance difference in the interface between the liver and the kidney (Morrison's space), and only a few waves are reflected back to the probe. Because of its high impedance, air interferes in the transmission of sound energy. Using gel between the probe and skin displaces air and creates a favorable interface for sound transmission.

As sound travels through tissue, energy is lost, which is a phenomenon known as attenuation. Attenuation is directly related to the impedance of the medium through which sound waves travel. The factors that determine attenuation are:

- refraction: the wave is deflected from its original direction when there is a change in terms of tissue composition between interfaces.
- scattering: the US bounces off in different directions after hitting a particle.
- absorption: US collides with particles and makes them vibrate, generating thermal energy and attenuating US power.
- attenuation occurs as sound waves travel from the US probe to reach its target, but also on its return back to the probe. The reflected waves are the returning energy that will be interpreted as the US image (Fig. 2). All these concepts are useful to understand time-gain compensation (TGC), a function that will be explained later [5].
- reflection: the main principle of the medical use of US, through which the wave bounces back to the source.

Artifacts

Different elements (e.g., air) can create artifacts that do not correlate with anatomy and may interfere with the interpretation of the images; however, in some situations, these artifacts provide information of the characteristics of the targeted tissue [6]. The following are examples of US artifacts:

-*Reverberation* occurs when the US bounces back and forth between two highly reflective interfaces, or between the probe and an interface. The probe receives these bouncing echoes later than the real returning waves and the image shows multiple lines (the brightness of the lines decreases with depth) with a similar distance among them. In this scenario, only the first line (closer to the probe) should be considered as real. Reverberation might be caused by air contained in usual or unusual locations: bowel gas, pneumobilia, pneumoperitoneum.

-*Ring-down or comet tail* is a type of the reverberation artifacts (Fig. 3). In this situation, the wave strikes against a small particle (such as a gas bubble) and resonates back to the probe at the same frequency after the original wave. In the comet tail artifact, the interfaces are so close to each other that the probe does not identify individual echoes. The resulting image shows a cone with the base close to the reflecting structure. Examples of structures that could explain the comet tail artifact are cholesterol gallstones or colloid cysts in the thyroid. The ring-down artifact is produced due to the reverberation between air bubbles. In this case, a highly echogenic line appears distal to the air bubbles as a series of parallel lines. This may arise when air is present within structures like the bile ducts or the portal veins.

- *Enhancement* of the intensity of the echoes between neighboring tissues is another common artifact. This is particularly useful to identify certain architectures, such as differences between liquid and solid structures. Examples of this artifact are cysts in solid organs, the gallbladder, and fluid collections (Fig. 4).

- *Acoustic shadowing* occurs when there is a lack of brightness, i.e. absence of image, distal to a high impedance structure that weakens the US. This phenomenon can help identify structures like gallstones, tissue calcifications, and bone (Fig. 5).

- *A mirror-image* can be seen when part of the US is reflected in a strong interface as it returns to the probe. The transducer interprets that the second signal originated deeper in the tissue, and it is shown as a second image underneath the high reflection structure. Examples of this artifact are liver tumors located close to the diaphragm, that appear in US images as two tumors equidistant to the diaphragm on both sides.

Doppler effect

The Doppler effect was first described by the Austrian physicist Christian Doppler in 1842. It describes the change in frequency when a sound wave is reflected in an interface in movement. When the interface moves towards the probe, the time between the emission, reflection and reception shortens (frequency increases). The opposite occurs when the particle moves away from the probe (frequency decreases). The Doppler effect is particularly useful in medicine since it gives information about the presence, direction and speed of flow inside a vessel. If the probe is held perpendicular (90 degrees) to the vessel, the Doppler effect does not occur. For this reason, to take advantage of this phenomenon, an angle of 60 degrees or less is recommended. The Doppler effect is shown as a color image in blue and red scales; a red wave means that the moving interface is coming towards the probe, while blue represents a wave moving away from the probe (Fig. 6).

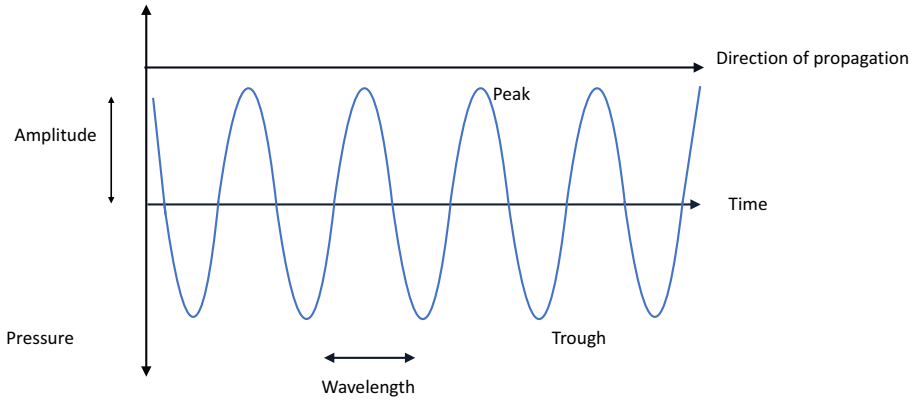

Fig. 1.. Graphic representing the physics of a wavelength, where all the variables involved are present. Description within the text.

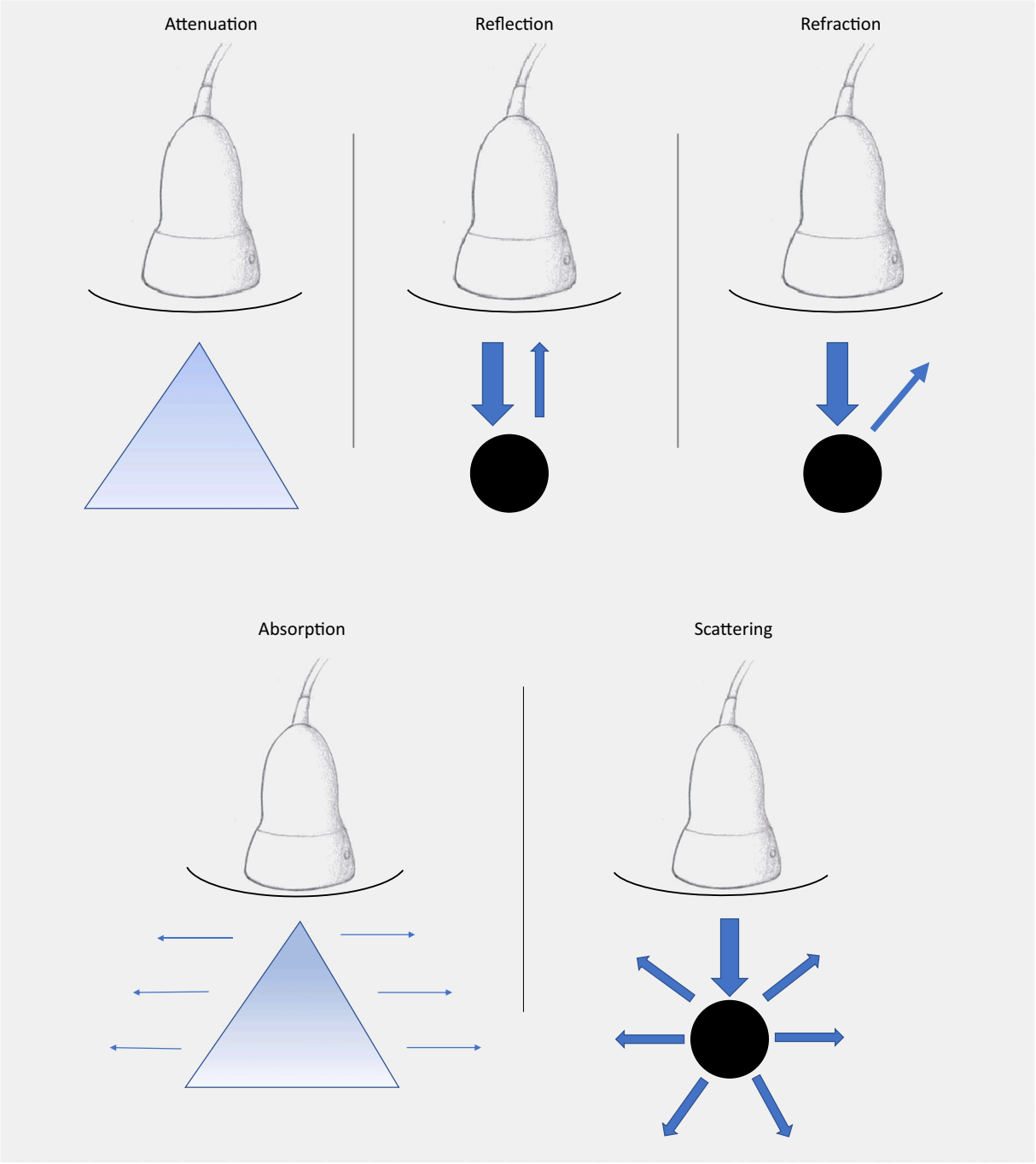

Fig. 2.. Different effects that end up in the attenuation of the images (seen as decreased light blue intensity within the footprint), including reflection, refraction, absorption, scattering, and descriptions are within the text.

Imaging modes

Different types of imaging modes are available in medical US devices: B, M, Doppler, and duplex imaging.The processor generates the images based on the information provided by the emission, reflection, and absorption of the US.

Even though the A mode (or amplitude modulation) is not available in US devices, it is easier to understand the other modes starting from the information provided by this basic one. The A mode represents the amplitude of a wave or echo in a timeframe, which is the most basic information one can expect from a cycle of US emissions.

In the B mode (or brightness modulation), the processor transforms the soundwaves into grayscale according to each echo's amplitude (A Mode). This allows for the generation of a bi-dimensional image that depicts human anatomy in real-time and produces the typical medical US image.

The M mode (motion) represents motion in a timeframe. This mode is particularly useful to assess deformation in anatomical structures in movement such as the diaphragm, cardiac valves and vascular structures.

The Doppler mode has been already explained, but, briefly, it is used to determine the presence of a fluid flow inside a vessel; it can also show the flow direction, speed and resistance.

The duplex imaging is the combination of the B and Doppler modes and identifies flow within a vessel, including its speed and direction (Fig. 6). If the quantification of flow characteristics is added to the

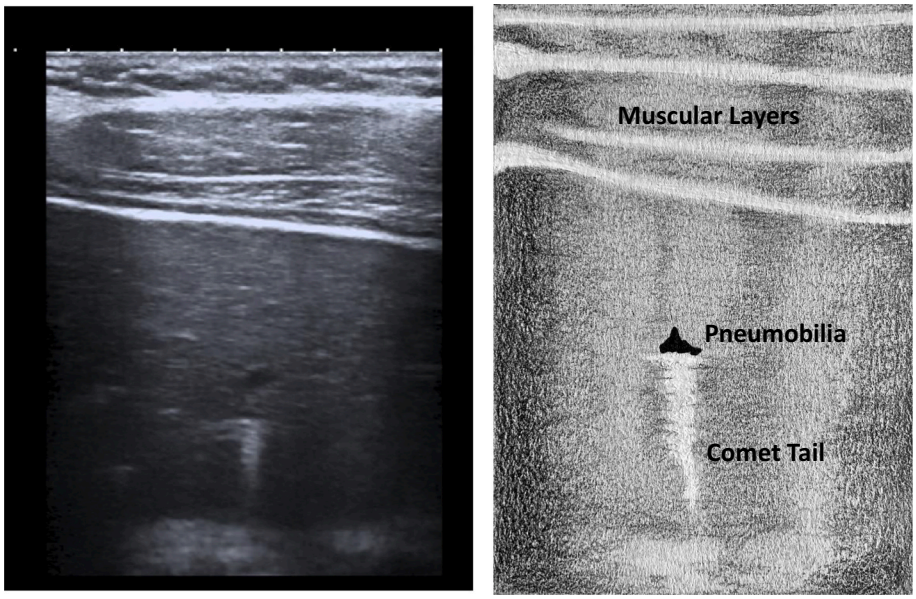

Fig. 3.. Pneumobilia creates an artifact called comet tail, where it can be observed thephenomenon as a comet tail, behind the air inside the biliary tract, (left image a current US image, right image a scheme to better understand the image to the left).

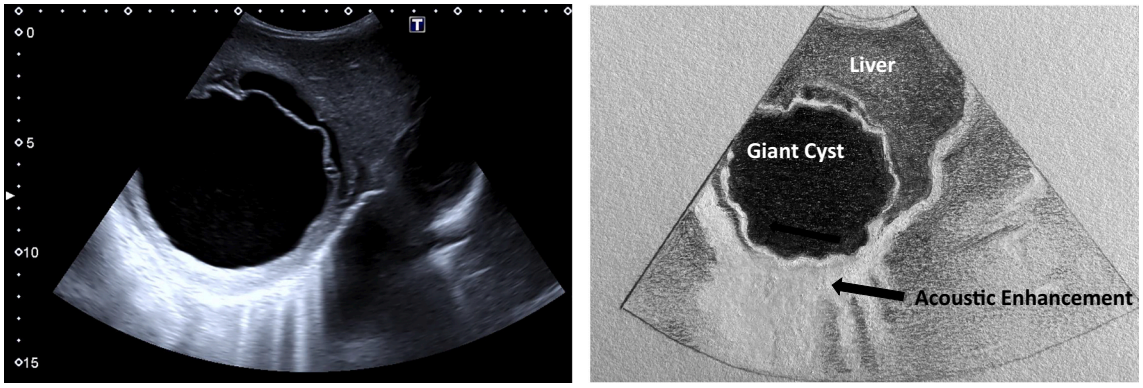

Fig. 4.. A giant cyst in the liver, actual image to the left, scheme image to the right.

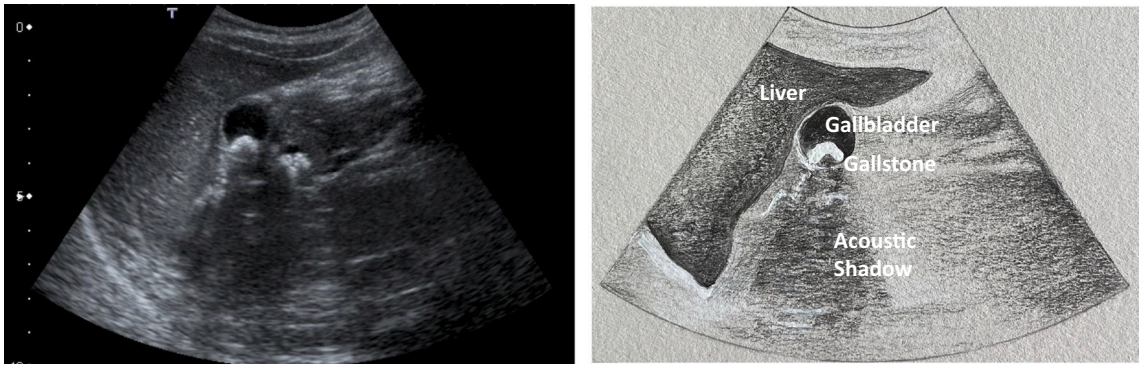

Fig. 5.. Gallbladder with stones. Acoustic shadows can be seen behind the stone.

duplex mode, a triplex mode image is obtained [7].

Safety

US is considered to be safe by the ALARA principle, which stands for "as low as reasonably achievable" and recommends avoiding exposure to radiation that does not have a direct benefit on patients. However, there are undesired effects that should be taken into consideration.

The oscillation of US waves generates a thermal phenomenon in tissues: it augments thermal energy with a consequent increase in the temperature in the interface. The amount of heat depends on the US wave frequency, the size of the scanned area, the time, and the

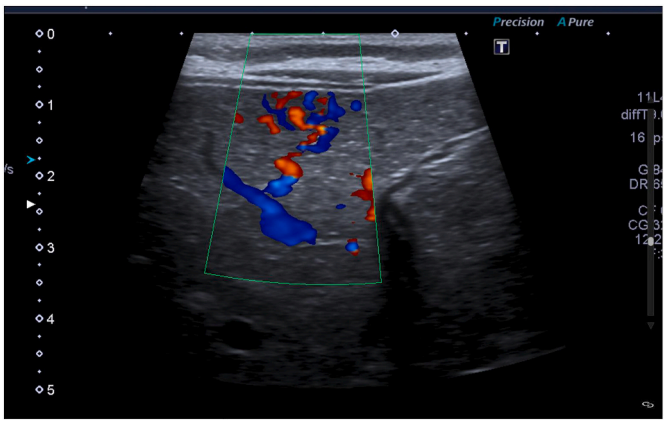

Fig. 6.. Doppler effect is used in a scan of the liver to assess the flows (portal, arterial) to differentiate the bile ducts and the vessels.

absorption capacity of the tissue. The higher the frequency, the more the temperature increases in the medium. Contrarily, a large scanned area allows for a cooling effect. However, modern devices do not generate a thermal rise greater than 1.5 °C above normal body temperature (37 °C), the maximum allowed by safety guidelines.

Another undesired effect is the mechanical consequence of particle vibration generated by US waves. This phenomenon produces micro-cavitations within the tissue, which could lead to modifications in membrane permeability. However, this effect does not have a significantly harmful effect on tissues.

Instrumentation

US devices comprise different components, all of which need to be carefully handled due to their fragility. Though US might seem a challenging technique to perform, it is an intuitive procedure and most of the devices share the same functions. In this regard, a complete understanding of the components and their functions will augment the chances of leveraging all the functionalities of the device [1,8].

A typical US device is composed of different parts: a generator, a monitor, a recorder and a transducer.

Generator

Briefly, the generator produces an electrical pulse ("power") that is transformed by the piezoelectric crystals located inside the transducer into sound energy (piezoelectric effect). The electrical energy is generated in pulses since a time gap is needed between emissions to "listen" or receive the returning US. The computer regulates the selected frequency, duration, and amplitude of the pulse. The higher the voltage, the brighter the image.

Once the US is reflected by the tissues, it returns to the computer through the transducer. The computer receives the echo, amplifies and filters it, and, finally, generates an image that is displayed on the monitor. The brightness of the image will depend on the intensity of the emitted and returning US. Assuming that the speed of sound is 1540 m/s, the time elapsed between the emission and the reception of the US will determine the distance from the object to the probe.

Transducer

As described earlier, the transducer is the key element of US scanning. It generates and receives the US energy, which is the information needed to produce the image. Transducers have evolved in the last three decades and, partly, this evolution is related to the necessity of different frequencies to perform a thorough scan. This led to the development of an array of transducers; those with low frequency, to increase the penetration of the US, and high-frequency transducers, to improve definition.

The selection of an adequate probe is important. There are 3 types of probes: linear, curvilinear, and phased array. The linear array has a scanning area with a rectangular shape (Fig. 3). The curvilinear exhibits a curvature that generates unparallel waves and a scanning area in the form of a truncated cone (Figs. 4, 5). The phased array probe has a small footprint with a wide base, which optimizes the view in the deepest part of the scanning area.

The selection of the frequency is as important as the selection of the format of the probe. Low frequencies (5 MHz) are suitable for the analysis of deeper organs, such as the kidneys, and high frequency probes (7.5–18 MHz) are used to scan superficial structures like the thyroid, vessels, muscles, etc.

Transducers can also be classified into transcutaneous (for example, TAUS), endocavitary (endovaginal, endorectal probes), or intra-operative open (IOUS) and laparoscopic (LAPUS). TAUS is a key tool for an initial approach to the abdominal cavity; but a more precise scan (CT-Scan or MR) is usually needed. Endocavitary probes have higher frequencies because of the proximity of the target tissue/organ. IOUS and LAPUS are mandatory in a hepato-pancreato-biliary surgery since they help determine the limits of tumors and their relationships with other vascular/biliary structures. Thus, US is a key tool for the assessment of resectability during surgical procedures.

In modern devices, two or three probes can be connected at the same time in different slots, so the selected probe can be modified from the keyboard without actually changing the connection to the device. Usually, during a TAUS, a curvilinear 3.5–5 MHz probe is needed. However, in slim patients, especially when scanning superficial structures (such as hernias or the appendix), a linear probe with a 7.5–10 MHz frequency range should be used.

Controls

Even though all devices are different according to the models and manufacturers, the basic controls are available in almost all units. Some features of the most widely available devices are described below.

Presets

In general, devices have different preset configurations. When a probe is connected to the machine, it automatically recognizes the type of transducer and suggests the sets that are suitable for that probe. The preset determines the frequency, gain, focus, depth, power, time gain compensation (TGC) and greyscale. The most frequent presets available in US devices are abdominal, vascular, soft tissue, neck/thyroid and obstetrics. There are also customizable presets that can be adjusted by the operator.

Gain/time gain compensation (TGC)

The returning echoes are attenuated by absorption, refraction and scattering. It is also known that the emitted ultrasound wave amplitude decreases as it penetrates tissue (a phenomenon previously described as attenuation). The receiver, through the "gain" function, can amplify these returning echoes to overcome tissue attenuation. But it should be considered that increasing the gain augments the overall brightness of the image, and excessive gain may result in increased image "noise" as all return signals are amplified. Sound waves returning from deeper tissues will have a greater attenuation. If the ultrasound image was constructed with the raw returned echoes, the image would be lighter in superficial layers and darker in deep layers, creating a confusing artifact. Most modern machines are equipped with a feature known as time gain compensation (TGC), or depth gain control, which allows users to selectively amplify the gain on signals returning from greater depths without increasing the overall noise in the closer field. Some machines have an automatic gain control function, which detects the decrease in echo amplitude with depth and applies compensatory amplification to

them. The function can also be applied manually by the operator from the keypad.

Frequency

Based on the tissue to be analyzed, the approach (intraoperative, transabdominal, or laparoscopic), the patient's anatomical characteristics, and the type of probe being used, it is imperative to preselect the desired frequency.

Depth

There is a timelapse from the emission of the US wave to the reception by the probe that is called the "go-return time". This parameter is useful to determine the distance or depth of the surface where the sound was reflected, assuming that the speed of sound is constant (1.54 m per second in soft tissues).

The ultrasound control "depth" establishes the distance of the structures in study. To some extent, the depth range is determined by the frequency of the transducer. For example, high-frequency (10–15 MHz) transducers generally cannot image structures located at a depth of over 10–15 cm. Similarly, lower frequency transducers (2–5 MHz) cannot properly image surface structures within the first few centimeters. Once the appropriate transducer is selected, the depth can be further adjusted to ensure proper examination of the area of interest.

Focus

The focus or focal depth is defined as the point on the beam axis where the beam intensity is maximum and its width is minimum. Just before the beam diverges there is an area known as the "focal zone". This area can be modified to correct the beam divergence and, for example, see more clearly the deeper parts of the screen. A strongly focused transducer has a narrow beam at the focus and a short focal zone with a fast beam divergence in the far field. A weakly focused transducer has a longer focal zone, and diverges less rapidly. This setting is also used in other imaging modes, such as Doppler, and is not exclusive to Modes "B" or "2D" [1].

Summary

Ultrasound is a non-ionizing radiation tool that can be used to acquire real-time images of different tissues and organs within the body. The components of a typical US device include the generator/receptor, monitor, recorder, and transducer. The generator produces a voltage pulse that is applied to the piezoelectric crystals inside the transducer or probe, which transform electricity into US waves. The computer regulates the frequency, duration, and amplitude of the pulse, and during the gap between pulses, the returning/reflected US is received. Once the US is reflected by the tissues, it returns to the computer through the transducer. The computer receives the echo, amplifies and filters it, and, finally, transforms it into an image that is displayed on the monitor. The monitor shows a real-time two dimensional image in grayscale. Using Doppler, colored images can be produced according to the direction of the flow (red towards the probe; blue away from the probe).

The transducer is the key element of US scanning, generating and receiving the wave that will produce the image. Different types of probes are available, including linear, curvilinear, and phased array probes, with different scanning areas and shapes. The frequency of the wave emitted by the probe is also important, with low frequencies being more suitable to image deeper organs and high frequencies, to image superficial structures.

It is important to understand the basic controls. Many US systems have preset controls, but the surgeon should master adjusting the controls, such as frequency, gain, TGC, depth and focus to optimize the US image.

US is a non-invasive, non-expensive, and real-time diagnostic tool that can be used in different medical specialties. It has become fundamental in hepatopancreatobiliary surgery for the evaluation and management of liver, pancreatic, or biliary tumors. Its physical principles and instrumentation need to be understood by surgeons in charge of performing it. The frequency, depth, gain, focus, and TGC knobs need to be adjusted according to the purpose of the exam to obtain the best possible image quality.

Training in US performance and interpretation should be part of the surgical residency program, to improve the surgical decision-making process, the perioperative management and, as a result, patient outcomes.

Disclosures

Catalina Poggi and Martin Palavecino have equally contributed to writing the present manuscript.

Catalina Poggi and Martin Palavecino have no conflict of interest.

No funding was required to write the paper.

Ethical approval statement

No ethical approval was needed to publish the present manuscript.

CRediT authorship contribution statement

Catalina Poggi: Writing – review & editing, Writing – original draft, Conceptualization. **Martin Palavecino:** Writing – original draft, Supervision, Investigation, Conceptualization.

Declaration of competing interest

None.

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