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Ultrasound Vascular Imaging (UVI)

1 Basic Principles of Ultrasound Sonography

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Introduction

Ultrasound imaging was first applied for medical purposes in the 1940s by Dr. Karl Theodore Dussik [1], and since then, it has spread all over the world as one of many non-invasive techniques by which we can gather information about tissues, cavities, and blood flow in the human body. It is an old technique but has a good cost–benefit ratio.

The main aim of this chapter is to provide a brief overview of the physics and instrumentation of ultrasound and Doppler imaging.

Sound

The main basic instrument of ultrasound is the sound itself. We can define sound as a sinusoidal wave that is an oscillation of pressures, transmitted through different tissues. This implies that sound mechanical energy will make a physical displacement of molecules through its passage, causing different types of rarefaction and compression in medium. In Figure 1.1, it is possible to see a sinusoidal wave, where the distance between two peak points is the wavelength (λ), the time corresponding to a complete cycle is a period (T), and frequency (f) is the number of complete cycles per second (f = 1 / T). The frequency of ultrasounds used in medicine ranges between 1 and 20 MHz, where hertz (Hz) corresponds to a cycle per second.

The velocity of sound in the medium is given by the equation $c = f \times \lambda$, and it is determined by the density and stiffness of the medium, being directly proportional to both. Each tissue has its constant propagation velocity, and does not change by wavelength or frequency of the emitted sound. Medical ultrasound devices assume an average propagation velocity of 1540 m/s [2, 3, 4, 5].

These concepts are extremely important because we are going to collect information about the depth and size of structures by using the principles of echo ranging and doing some calculus. In ultrasound, in order to obtain distance, this equation is used: $d = c \times t / 2$, where d is distance, c is propagation velocity, and t is the time that the ultrasound signal takes to make a round trip from the transmitter, reaching the target and coming back to the receiver (Figure 1.2).

After applying an ultrasound pulse, the time that echo returning takes is measured, and, when the propagation velocity sound for that tissue is known, the depth of the interface is calculated. This measurement is influenced not only by the accuracy of the value that is given for propagation velocity from tissue, but also by the path that sound pulses travel.

Sound amplitude is represented by the height of the wave and can be described in decibels (dB), which do not represent an absolute signal level but a logarithmic ratio of two amplitudes (A2 and A1). By halving the amplitude, the decibel's level goes up by six ($dB = 20 \log (A2 / A1)$).

Medium

In clinical use, ultrasound images are formed by echoes (reflected sound), and to do this, there must be tissue with reflecting capability. If the medium is totally homogeneous, there won't be a reflected sound, so it will look like a cyst – in other words, an anechoic medium. At junctions of tissues with different properties, the sound will reflect, and different intensities of echoes will return. The amount of reflection (or backscatter) is called acoustic impedance (Z), and it is determined by tissue density (ρ) and the propagation velocity of the tissue (c) ($Z = \rho \times c$). These reflections are determined by the differences between the values of each tissue's acoustic impedance. This means that a big difference in a tissue's acoustic impedance (like bone and air) will reflect almost all sounds, and smaller ones (between fat and muscle) will reflect only some sound while the other

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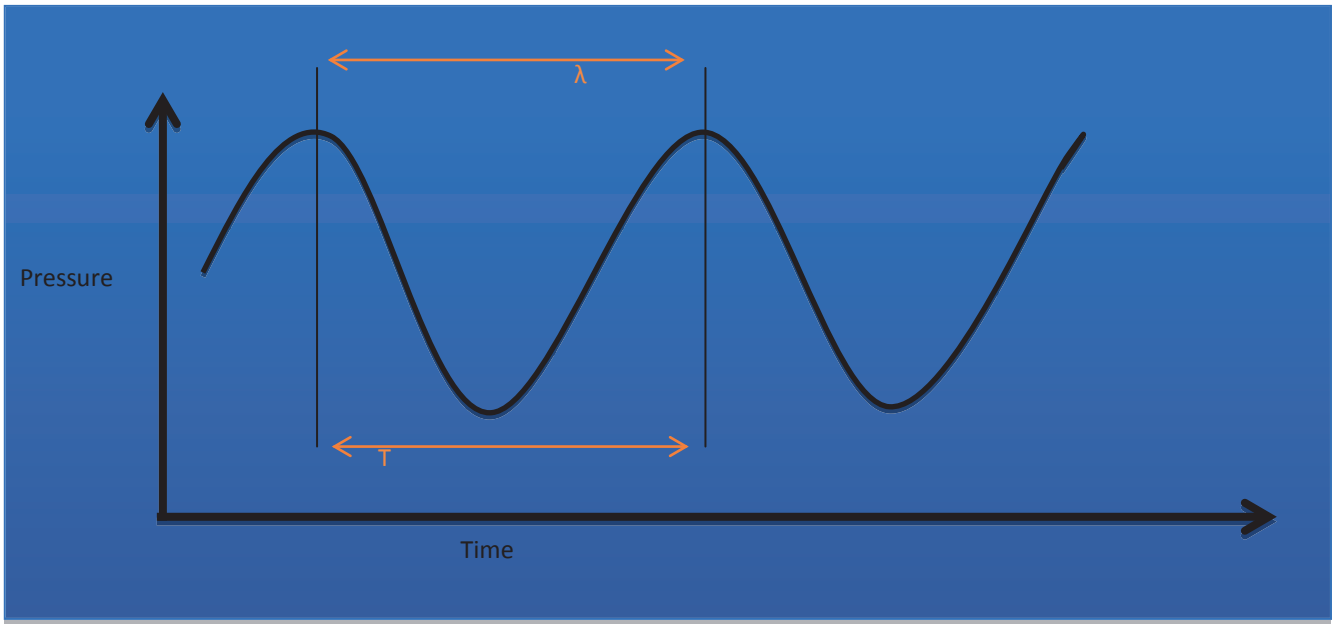

Figure 1.1 Sound wave. The graphic represents oscillating pressure producing compression and rarefaction of the medium. λ: wavelength, the distance between two peaks in the sine wave; T: period, the time of one cycle.

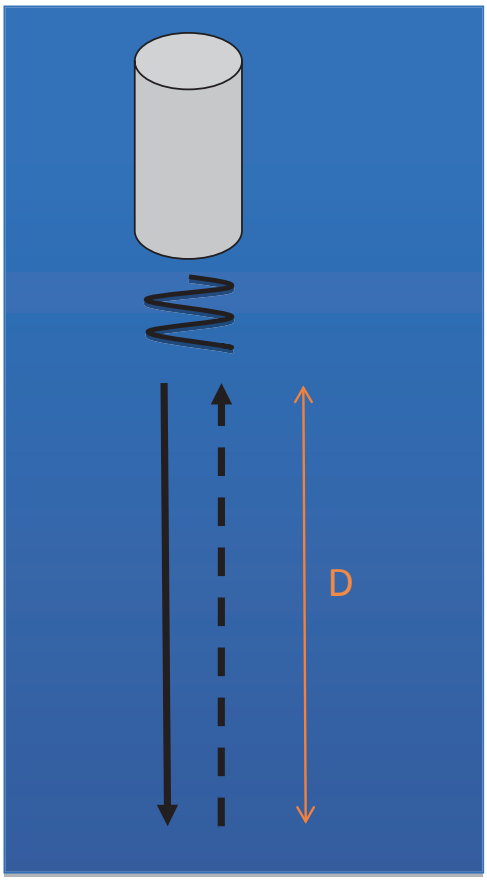

Figure 1.2 Ultrasound detection. An impulse wave is started by the transducer, goes through the medium, reaches the desired target and is reflected until the transducer. D: Distance between transducer and target.

amount will continue forward through the other tissues ahead. Therefore, it is possible to say that ultrasound does not give images of tissue structures but, alternatively, interfaces between tissues with different levels of acoustic impedance.

Interface of reflection

There are two types of reflection interfaces, specular reflectors and diffuse reflectors. The first one acts like a flat mirror, and the second one acts like a mirror ball (i.e., a "disco ball"). This happens because of the surface and size of the interface: if it is large and smooth (like a diaphragm or the wall of a urine-filled bladder), it will be a specular reflector, whereas if it is small and wrinkled (like the liver parenchyma), it will be a diffuse reflector. The first one will reflect, and the second one will scatter (Figure 1.3).

The reflection coefficient (R) is the energy reflected as a fraction of the incident energy, where Z_1 is proximal tissue impedance and Z_2 distal tissue impedance ($R = (Z_2 - Z_1)^2 / (Z_2 + Z_1)^2$).

Because the specular reflector acts as a flat mirror, the returned echo depends on the degree of insonation. If the angle is too far away from a 90° angle, the sound will reflect and won't be detected by the transducer. This is very important because the ultrasound image is reconstructed only by the reflections that return to the transducer, so if the echo does not return, there won't be a good interface for images. In fact, most of the echoes come from the diffuse reflector.

These interfaces produce a speckle on the ultrasound, resulting from a constructive and destructive interference of the reflected sound, directing away almost all energy with only a small part reflected toward the ultrasound transducer. Because of this property, the ultrasound incident angle does not change the amount of echo recognized (Figure 1.4). Thus, when analyzing vessels, if the wall is the main purpose of the study, it should be insonated at a 90° angle; otherwise, when studying flow in Doppler imaging, this angle should be less.

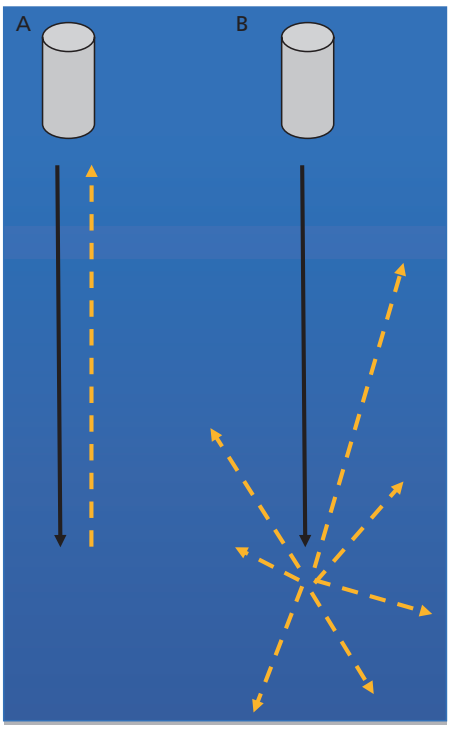

Figure 1.3 Specular (A) and diffuse (B) reflectors. In specular reflectors, the sound is reflected by a "flat mirror" coming straight to the transducer, if the beam was at a 90° angle. In diffuse reflectors, the sound is scatter in all direction, returning just a small amount to the transducer.

Refraction is the alteration trajectory of ultrasound wave when it passes through different tissues with different acoustic propagation velocity. This means that not all echoes received in a transducer come from a straight line; they could come from a different depth or location, changing the liability of the image displayed (Figure 1.5). To reduce this artifact, the emitted ultrasound angle should be (as much as possible) perpendicular to the interface under study. Snell's law is the equation that relates the angle of refraction and propagation velocity ($\mathrm{Sin}\,\theta_1 / \sin\theta_2 = c_1 / c_2$).

If we think of sound as an acoustic energy, while it passes through different interfaces, it will decrease wave amplitude by transferring some of that energy as heat to the ambient surrounding it – absorption – and it will reflect and scatter the rest.

The amount of energy produced in time is called acoustic power (expressed in watts [W]); and, when the energy is related in spatial distribution, it is called intensity (I) ($I = W / cm^2$).

So attenuation is the medium capacity of decreasing ultrasound signal amplitude, which means that determines the ultrasound penetration efficiency. It is measured in decibels (dB), allowing one to compare different levels of ultrasound intensity, and is calculated by this formula: $dB = 10 \times \log_{10} I$.

Attenuation is a result of the combined effects of absorption, scattering, and reflection, and it depends not only on the property of the medium but also on the insonating frequency. Higher emitted frequencies result in less penetration because attenuation is more pronounced:

$$\text{Attenuation} = \alpha \times L \times f$$

where α is the attenuation coefficient of a tissue, L is medium length, and f is frequency.

Greater penetrations depths are achieved by using transducers with lower transmit frequencies or increasing

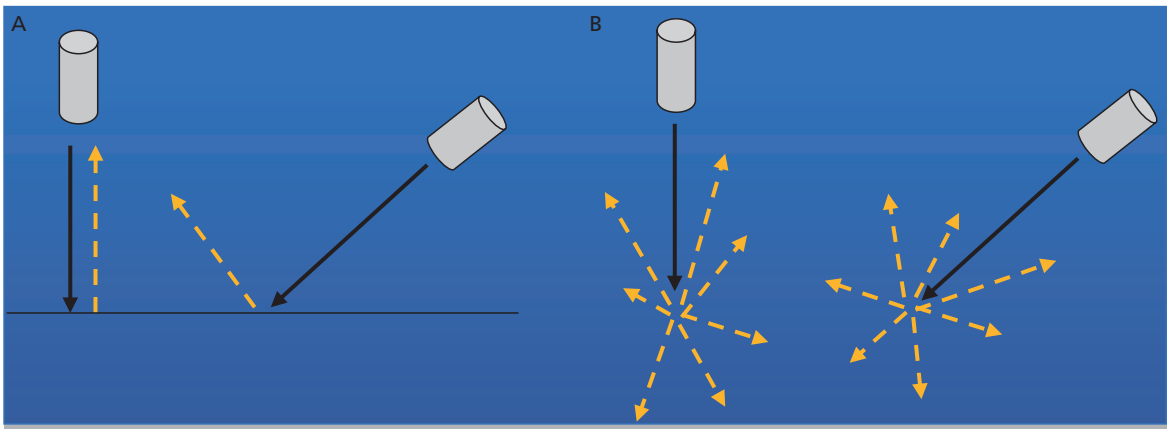

Figure 1.4 The incidence beam and the type of reflector.

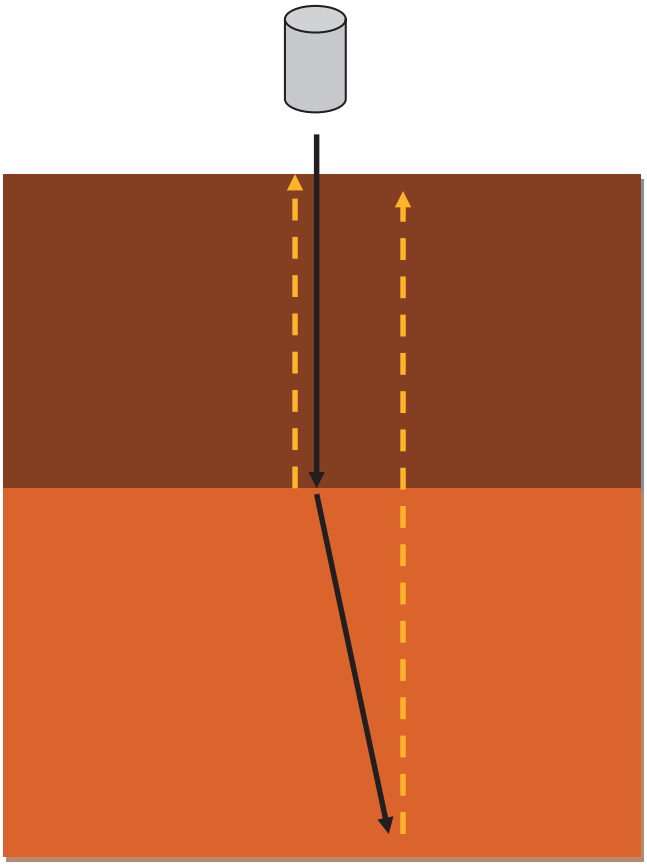

Figure 1.5 Different tissues have different propagation velocities, leading to changes in the direction of the sound wave – refraction.

amplification of the signal. This amplification is known as time-gain compensation or depth-gain compensation. The gain is chosen according to the depth and localization of the study vessel, and it is crucial for acquiring a signal with good amplitude and intensity. Together with the output energy and signal-to-noise limit, it is a parameter of great importance to be managed by the operator.

Doppler ultrasound

The study of blood vessels is based on the Doppler effect, which was discovered by Christian Doppler in 1842 [4, 5]. When applying a high-frequency beam into a moving tissue, instead of being reflected with the same frequency (as if it was a stationary interface), its frequency of reflection will change and will be directly proportional to the velocity of the interface, increasing when it is coming closer and decreasing when moving farther away (Figure 1.6). In our body, the moving red blood cells within vessels are the main cause of the Doppler shift. Microbubbles, breathing, heartbeats, bowel peristalsis, and movements of the transducer also produce this effect.

The Doppler frequency shift (f_d) must be understood as the difference between the frequency of sound reflected from the moving target (f_r) and the frequency of the ultrasound beam transmitted into tissue (f_t) ($f_d = f_r - f_t$).

The frequency of sound emitted can be selected by the user, but the frequency of sound reflected (echo) depends not only on the type of transmitted sound but also on four other factors: the direction and velocity of the moving target (v), the velocity of sound in the medium (c), and the angle of the ultrasound beam (θ):

$$f_d = 2 \times f_t \times v \times \cos\theta / c$$

To fully detect the frequency shift, the angle of the beam should be 0°. Above 60°, the cosine of that angle changes abruptly, causing difficulties in estimating correctly the Doppler shift. If a vessel is insonorated at a 90° angle, the cosine of θ will be zero, at which point there is no Doppler frequency shift. Between 0° and 60°, there are losses of almost 50% of shift detection, so it is in our best interest to use Doppler angles less than 60° to estimate velocity more correctly (Figure 1.7). It is possible to increase Doppler shift by diminishing the beam angle.

In summary, the best vessel wall imaging is given when it is used at a 90° angle, between the transducer and the wall, while the optimal Doppler frequency shift is acquired when the transducer and the direction of flow have a small angle or none at all.

Doppler frequency shifts are within the range of human hearing [4–6], so they can be heard at the same time as they are displayed in a graphic of time versus frequency spectrum data. Therefore, when doing the Doppler, a trained operator should hear carefully the sound flow characteristics together with the information coming from the Doppler frequency spectrum display.

Instrumentation

In Doppler studies, the main purpose is to obtain information about flow. This can be achieved by using Doppler devices with continuous-wave or pulsed-wave ultrasound. The first is the simplest one, and it is used at bedside or intraoperatively to verify the presence of flow in superficial vessels. The system has two transducers that transmit and receive ultrasound continuously. The main limitation of this type of Doppler is its poor capability to discriminate among different depth signals. This happens because the transmitted and received beams have some overlapping areas, causing difficulty in detecting with certainty the source of the signal. Therefore, continuous-wave Doppler can determine flow direction, but it is not the best choice for studying blood flow accurately.

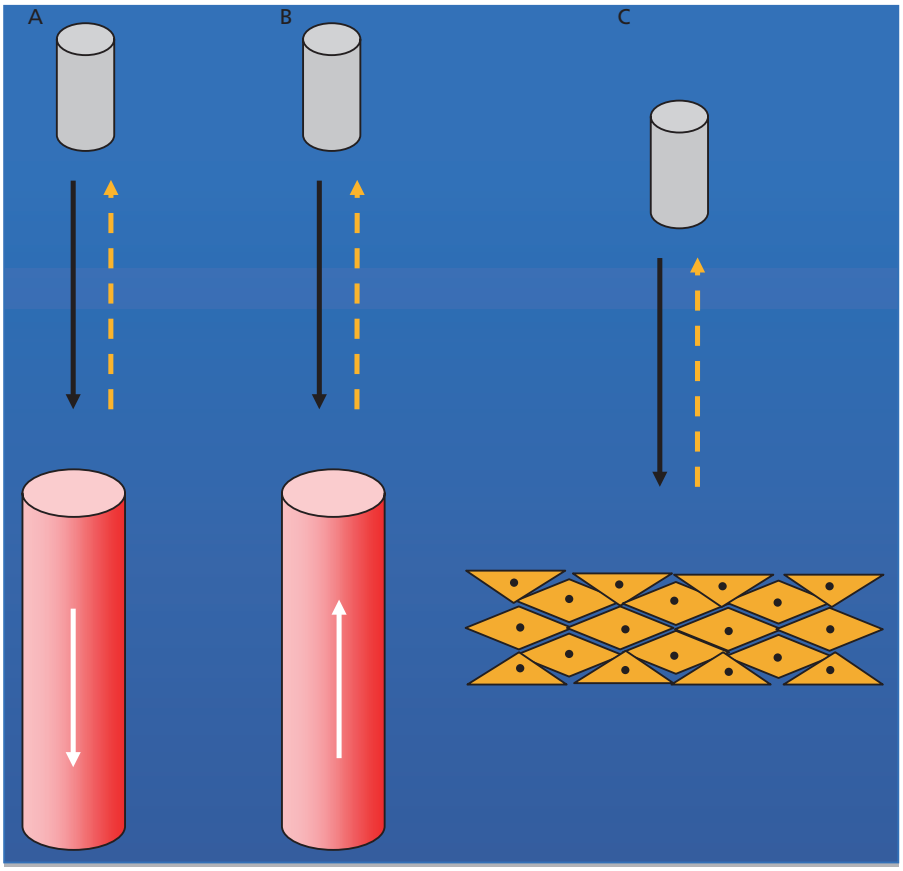

Figure 1.6 Doppler effect on moving (A and B) and stationary targets (C). When the moving target is coming away from the transducer (A), the difference between the reflected and the transmitted frequency is less than zero. The opposite occurs when the moving target is getting closer to the transducer, the received frequency is bigger than the transmitted one, and the Doppler frequency shift becomes positive (B). In stationary targets (C), the backscattered ultrasound has the same frequency as the transmitted sound, being zero in the Doppler shift.

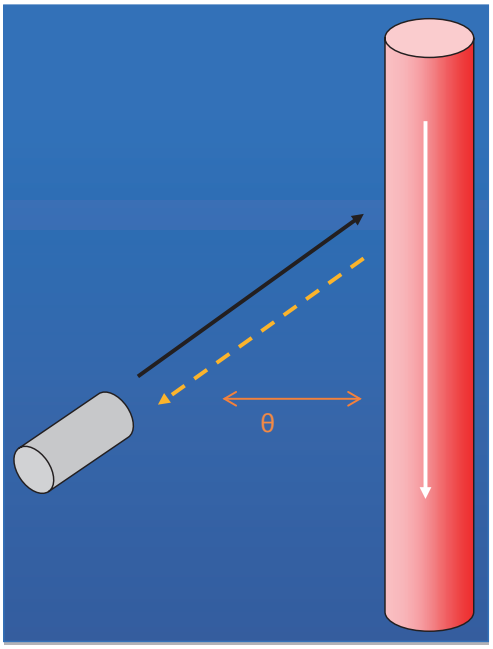

Figure 1.7 The Doppler shift depends on the velocity and direction of the moving target as well on the Doppler angle (θ) – the angle between the transmitted beam and the vessel.

On the other hand, pulsed-wave Doppler gives information with more accuracy and fewer constraints. Instead of a continuous emission of ultrasound waves, this type of Doppler study uses an emission of brief pulses of ultrasound, recording the reflected signal in between (Figure 1.8). By knowing the depth of the vessel that it attempts to study, one can choose the time interval between the transmission and the return of the ultrasound, thus obtaining the Doppler shift information of that vessel only. Pulsed-wave Doppler is normally combined with real-time grayscale ultrasound imaging, forming a duplex Doppler ultrasound. Duplex ultrasound provides flow information from a sample volume at a defined depth, allowing the calculation of blood flow velocity through the angle of incidence of the beam and the vessel.

Doppler display

The Doppler display can have two types of images: the Doppler frequency spectrum waveform and Doppler imaging (Figure 1.9).

The first type, Doppler frequency spectrum waveform, is a graphic that has acquired variations in the time of Doppler

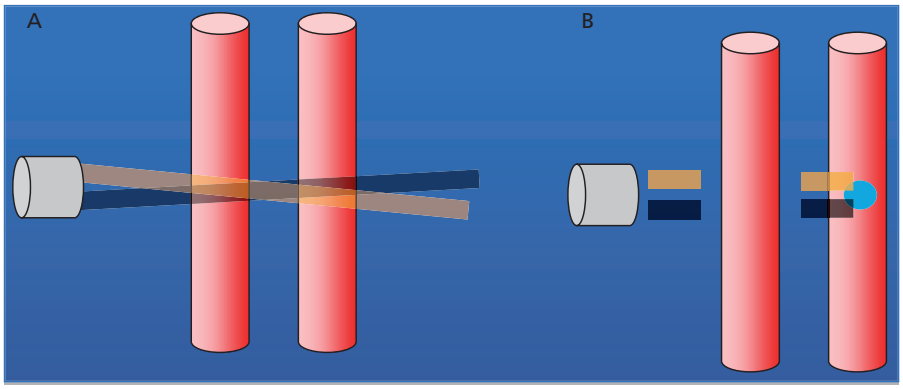

Figure 1.8 Continuous-wave (A) and pulsed-wave ultrasound (B). Continuous-wave ultrasound has two transducers that transmit and receive ultrasound continuously. Here, the transmitted and received beams have some overlapping areas. Pulsed-wave ultrasound emits brief pulses of ultrasound, recording the chosen reflected signal, obtaining the Doppler shift information only of the target of study.

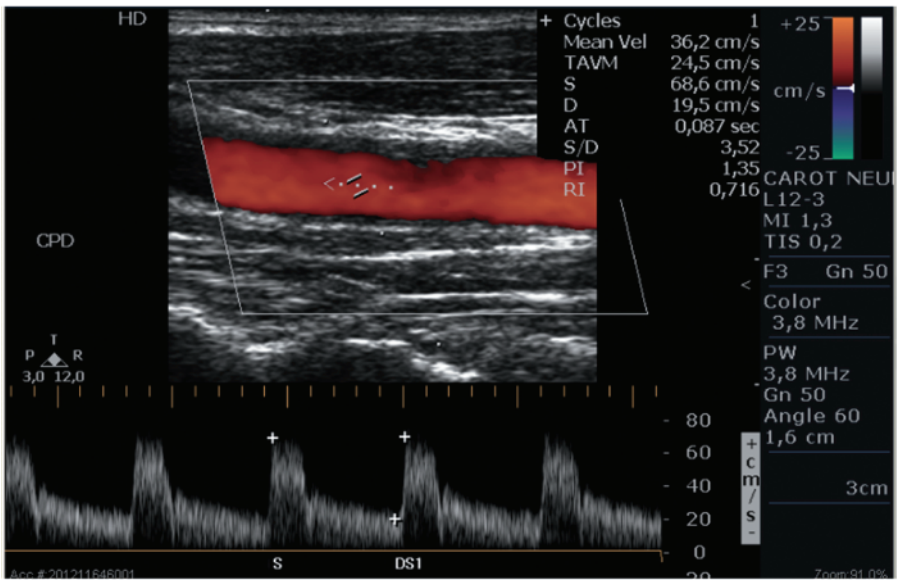

Figure 1.9 Doppler spectral display. Doppler spectrum of the common carotid artery longitudinal plane with optimal sample volume with flow above the zero baseline.

frequencies. "Spectrum envelope" is the name given to the maximum frequencies at a certain point of time, "spectrum width" indicates the range of frequencies at a certain point of time, and "spectral broadening" results from the presence of a large number of different frequencies at a certain point of time.

In the second type of display, instead of having a graphic, information is displayed as Doppler imaging. Here, it is possible to receive and see information about a stationary target (e.g., the wall of the vessel) and a moving target (e.g., red blood cells). The presence and direction of motion are given by the signal phase, and the velocity of the motion is given by changes in frequency. Different colors indicate different directions of movement, and different degrees of color saturation indicate each frequency shift.

Doppler spectral display

Detection of a Doppler frequency shift implies the movement of a target along the beam axis, and their positive or negative values indicate the direction of the movement. A Doppler spectrum is constructed using fast Fourier transform. Through this equation, all Doppler shifts acquired in the sample volume are individualized and displayed as a function of time. In Figure 1.9, the x-axis is time in seconds, and the y-axis is flow velocity in m/s or cm/s, or can be switched to display Doppler frequency shift measured in KHz. Brightness of pixels corresponds to the quantity of red blood cells moving with the same velocity in a certain instant of time. If the flow has its movement against the Doppler beam, it will be seen in the positive part of the y-axis (normally above the zero baseline); if the flow has the same

direction as the Doppler beam, it will appear below the zero baseline.

The cardiac cycle is an important mechanism that influences spectral waveform. During systole, there is the highest flow velocity, increasing the Doppler shift values, while in diastole, flow velocity decreases, diminishing Doppler shift values.

Pathological abnormalities can be seen and interpreted when analyzing these graphics. In vessels stenosis, normally there are large Doppler frequency shifts in both systole and diastole at the place of stenosis. In vessel vasodilation, there is increased systolic amplitude and a rapid flow through diastole.

The systolic-diastolic ratio, resistive index, and pulsatility index are the Doppler indices also studied in spectral Doppler. Variables such as cardiac cycle, resistance to flow, blood pressure, vessel wall length and elasticity, extrinsic organ compression, hematologic factors, and other factors influence the Doppler indices and the other measurements of peak systolic and end-diastolic velocities. So, it is very important to be aware of these factors when interpreting the Doppler spectrum.

Color Doppler imaging

Color Doppler, or color-coded duplex ultrasound, detects the mean Doppler shift within the sample area, superimposing a color upon the grayscale image (Figure 1.10). The signal from flow is displayed in a color scale, determined by the direction of its movement. The relative frequency shift obtained by the red blood cells' movement determines the degree of saturation of the color displayed. This is important because it allows an estimation of relative velocity from the image itself.

Color Doppler imaging permits one to identify, inside of the vessel, small and localized areas of turbulence, which aids in the diagnosis of stenosis or irregularities from the vessel wall. This study has more sensibility than duplex study, as it enhances areas of turbulent flow that otherwise wouldn't be seen.

The color Doppler sample volume is a box, and only the tissue within it will be analyzed. A small box allows a better gathering of information. The color map is localized near the image to indicate the colors used, and it is divided in the middle by a black bar, which corresponds to the zero-flow point on the scale. The color above this baseline, normally red, shows flow toward the transducer, and the color below the baseline, normally blue, shows flow away from the transducer. Brighter color indicates the highest mean velocities, and dark colors indicate lower velocities. In some instruments, different color shades or even different colors can be used for higher and lower velocities. The numerical scales in these maps indicate mean frequency shift, and they must be adjusted to the type of vessel that is under study.

It is important to understand that color Doppler doesn't show true velocity; it only indicates the weighted mean frequency shift measured in the vessel. Any velocity can be represented by any color under special circumstances. So flow velocity, the Doppler angle, aliasing, pulse repetition frequency, the color map used, and the cardiac cycle will influence the color displayed inside the blood vessels.

Color Doppler energy (power mode)

Instead of displaying color imaging through frequency information and its mean, here the image is obtained from integration of the power of the Doppler signal. In other words, the image isn't constructed with information about flow

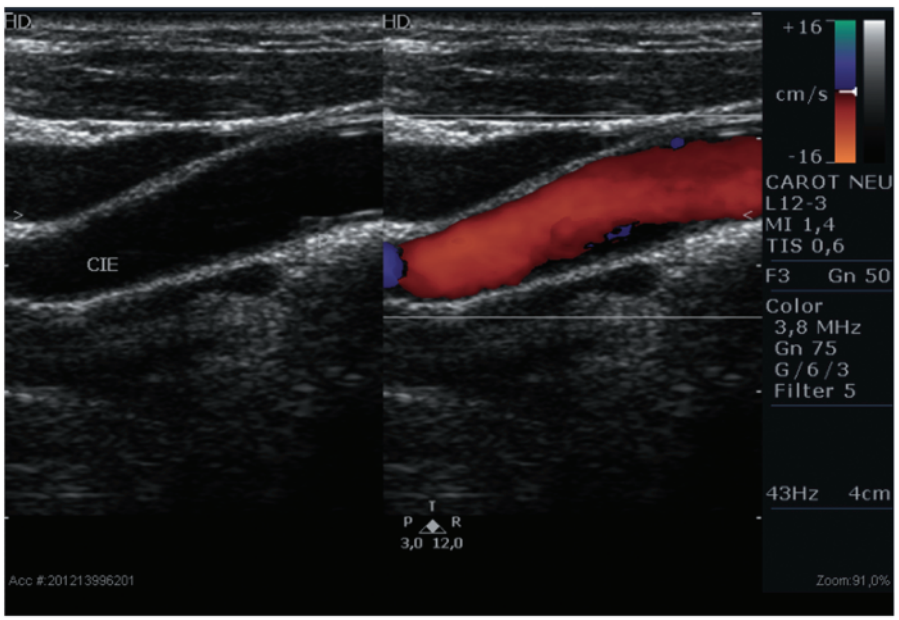

Figure 1.10 Color Doppler imaging. B-mode imaging and color mode imaging of the internal carotid artery.

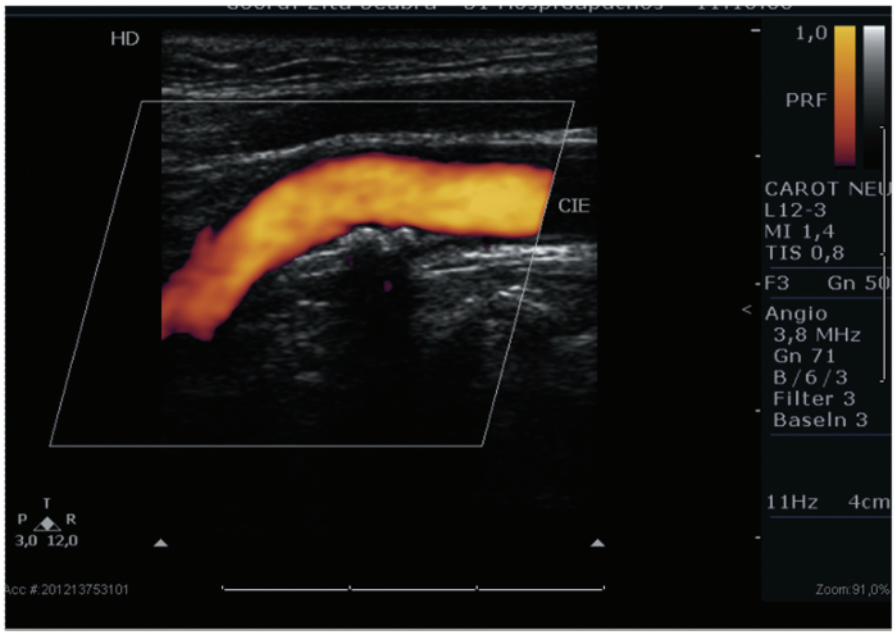

Figure 1.11 Color Doppler energy (power mode). Power mode imaging of the internal carotid artery.

direction or velocity; it just shows the distribution of the power and amplitude of the Doppler signal. This technique is used more in the evaluation of parenchymal flow and in studies of tumor vascularity. The sum of all signals is related to the number of moving red blood cells, and it is represented by levels of brightness. Information about flow direction or velocity isn't shown. The advantages of power Doppler are fewer artifacts from aliasing, a better signal-to-noise ratio, and the image is much less angle dependent, increasing the gain settings and improving sensitivity for flow detection. Some disadvantages are that there are more movement artifacts and no ability to show qualitative or quantitative information about flow velocities (Figure 1.11).

In summary, both spectral and color Doppler complement each other and should be used together. The first one display all Doppler shifts detected in a certain time, whereas color Doppler shows flow information using the mean of the Doppler shifts. Color Doppler imaging is used more to detect vessels and to confirm and determine the direction of blood flow, while spectral Doppler ensures a better characterization of this flow with more precise measurements [3, 5, 6].

Causes of artifacts

Frequency

The main aim of Doppler is to measure, in the most precise way, the blood flow and its main characteristics. It is achieved by the source signal coming through red blood cells, as they act like a scatter reflector. In order to select a Doppler frequency, it is important to know that the intensity of the return echo is proportional to the fourth power of the frequency (FREQ^4). With higher transducer frequencies, Doppler sensitivity is increased, but with the cost of penetration, that lowers because of higher attenuation. So the knowledge of the operator is very important in deciding whether sensitivity or penetration is more important for examination, and in balancing both.

Spectral broadening

The signal returned from a vessel is not of single frequency, but of a range of frequencies coming from different blood flow velocities. When there is a large range of flow velocities such as this, it is called "spectral broadening" and can indicate a site of turbulence flow and vessel narrowing. This broadening can be created by the operator – intrinsic spectral broadening – when he or she uses a transducer too close to the vessel, making the incidence beam too wide and causing multiple beam angles between the transducer and the vessel. It is assumed that there is only one Doppler angle, but in reality there are multiple angles, causing an increase in range of Doppler shift frequencies. Excessive gain, large sample volume selection, and placement of the sample volume near the vessel wall can also broaden the spectrum range.

Scattering and acoustic shadowing

These artifacts are produced by air and bone tissues that don't allow penetration of the ultrasound. This prevents the examiner from evaluating the structures behind these tissues. Pressing the transducer against the airy tissue (allowing the air to move to the sides) and positioning the transducer around the impenetrable tissue are two methods to overcome this issue.

Mirror artifact

This artifact occurs when tissues with great reflecting capability mimic structures, displaying a reflected image behind them, in grayscale imaging or in duplex imaging.

Aliasing

Aliasing is an artifact that occurs in pulsed-wave Doppler, when the pulses emitted have less than twice the frequency of the maximum frequency shift produced by blood flow. The Nyquist limit dictates that to measure a frequency correctly, it must be sampled at least two times per period (δF_{Max} = pulse repetition frequency / 2). If it sampled less often, aliasing will occur, reconstructing an artifactually lower frequency. So, the sampling rate of the transducer, or pulse repetition frequency (PRF), has to be twice or more the highest maximum frequency of the moving target. The PRF will determine the maximum Doppler shift frequency that is possible for accurate measurement of direction and amplitude blood flow ($V_{Max} = c / (4 \times T \times F_0 \times \cos\theta)$).

In spectral display, aliasing appears as a wraparound of the higher frequencies below the baseline in the display. In color Doppler display, this phenomenon also occurs, passing through a transition of unsaturated color from one flow direction to the opposite with wrapping around of the frequency color map. Increasing the PRF or Doppler angle, lowering the frequency of the Doppler transducer, and shifting the baseline in spectral display and color scale will reduce aliasing.

Doppler angle

To make accurate measurements of Doppler velocity, it is necessary to know the correct Doppler angle. Doppler angles should be, in most cases, below 60°, because small changes above this threshold will result in significant differences in the calculated velocity. Errors in angle calculation can result in a completely different velocity when it is used at higher angles than at lower ones.

Doppler indices measurements, such as those of the resistive index, are not influenced by Doppler angle, because they are based only on the relationship between systolic and diastolic amplitude.

Gain

Gain is essential for acquiring all information from the reflected echo and permitting an accurate measurement of Doppler velocities. The gain should be adjusted for B-mode, Doppler, and color duplex scanning. In Doppler with excessive gain, noise will appear and can result in overestimation of velocity. The opposite, insufficient gain, can result in underestimation of peak velocity. These types of phenomena are the main reasons for the poor reliability of planimetric measurement of stenosis based on the vessel's cross-section depicted in color duplex imaging.

Wall filters

The signal received from Doppler frequency shifts comes not only from blood flow but also from patient motion. This patient motion can be voluntary or involuntary (e.g., breathing or peristaltic movements) and normally has a low Doppler frequency shift. Wall filters are used to remove all frequencies below a predetermined threshold frequency. Operators should know these filters and configure them when analyzing vessels with a low blood flow, changing thresholds when necessary.

Hemodynamics

Blood is a very complex fluid, and vessel walls have some characteristics that make the physics of blood flow not as simple as it could be when applying laws of hemodynamics. These laws are applied to Newtonian fluids (solutions with constant viscosity, like water) when flowing in tubes, dictating that flow velocity is a function of the pressure difference between the two ends of the tube. Three factors determine the velocity of the fluid: viscosity, geometry of the vessel, and pressure.

$$Q = (P_2 - P_1) \times \pi \times r^4 / 8 \times l \times \eta$$

This equation, the Hagen–Poiseuille law, shows that the volume flow rate (Q) is proportional to the vessel diameter (r) and inversely proportional to its length (l) and the viscosity of the fluid (η). This equation is applied just for moving fluids through a cylindrical vessel.

Types of flow

Laminar flow

The velocity of blood is related to its position in the vessel and its viscosity. At the periphery of the vessel, the flow resistance is greater; therefore, blood flow velocity is lower. Coming into the vessel center, the velocity of blood becomes higher, reaching the maximum in the center. In perfect conditions, without turbulent flow and with neither the narrowing of vessel diameter nor changes in direction, laminar blood flow will have a parabolic shape.

Laminar flow is characterized by smooth and constant fluid motion, and it occurs when viscous forces are dominant (Figure 1.12). Reynolds number (Re), which is a dimensionless number, is calculated by the product of velocity (v) and mass density (p) divided by the viscosity of fluid (η) and multiplied by the diameter of the vessel (d). When this ratio is below 2000, flow is laminar; when it is higher, it is called turbulent flow, where the inertial forces are the dominant ones.

$$Re = v \times d \times p / \eta$$

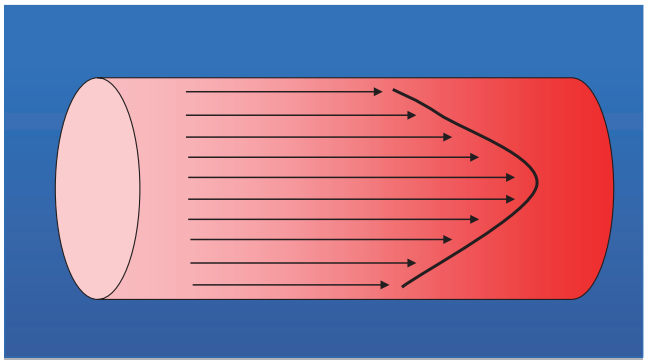

Figure 1.12 Laminar flow.

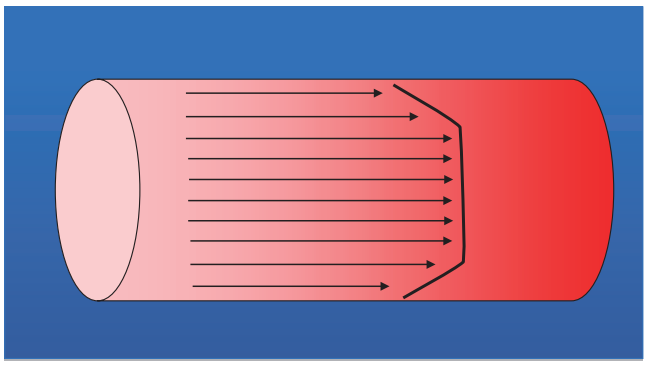

Figure 1.13 Plug flow.

In practice, this is the normal narrow spectrum, with a systolic window and decreasing velocities from the middle to the walls of vessels. The same happens with color Doppler, whereas brighter colors are seen in the middle, and dark ones toward the wall. During systole, the flow velocity reaches its peak values; on diastole, flow velocity decreases at the lowest point, and can even reverse direction in some conditions.

Plug flow

In the aorta and other large-diameter vessels, plug flow is the type of normal flow observed. Instead of just high-speed red blood cells in the vessel center, there is a wide band of them moving very fast, forming not a parabolic but a plug-shaped velocity profile. The Doppler spectrum is narrower than laminar flow, demonstrating a majority percentage of red blood cells moving with the same speed (Figure 1.13).

Disturbed blood flow

When red blood cells come across vessel bifurcations and stenosis, some change their directions without changing the direction of all blood flow. In these areas, flow velocity tends to increase, with higher systolic peak and a broadened spectrum.

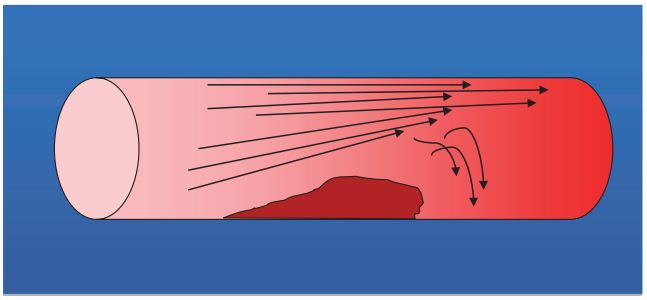

Figure 1.14 Turbulent flow.

Turbulent flow

This type of flow is visualized in high-grade stenosis and in high-velocity flow locations as shunts and fistulae. It is characterized by random and chaotic flow, without a precise direction of red blood cells (Figure 1.14).

In practice, there is a mosaic-flow pattern in color duplex ultrasound, severe spectral broadening, obliteration of systolic window, less pulsatility flow, concentration of flow with lower velocities, and formation of flow separation and eddy currents downstream the stenosis, where the blood vessel lumen widens.

Flow resistance

High-resistance blood vessels

This pattern is typical of peripheral flow, when small arteries and arterioles contract, increasing resistance to blood flow in order to maintain a high level of blood pressure. This results in a more pulsatile flow, allowing just a few flows to reach the capillary bed during diastole.

High-resistance flow is normally seen in skin and skeletal muscle arteries suppliers. The external carotid artery (ECA) can also show a high resistance pattern, with a typical triphasic pattern in the Doppler spectrum, a sharp and rapid rise of peak systolic velocity with a rapid fall after the end of systole and a very low velocity flow in diastole with the same direction as in systole. In the other high-resistance vessels, flow reverses direction in early diastole and is very low or nonexistent in the rest of diastole.

When necessary, arteries with this type of pattern flow can reduce their resistance, allowing more passage of blood, primarily during diastole [3]. This transition between high- to low-resistance flow can be physiologic (exercises) or pathologic (arteriovenous shunt).

Low-resistance blood vessels

This pattern is characterized by a forward blood flow in all cardiac cycles with mild systolic rise followed by a steady flow in diastole. This happens mainly in arteries of paren-

chymal organs and in the brain, where it is vital to the continuous supply of oxygen.

Instead of a triphasic waveform like in high-resistance vessels, here there is a biphasic pattern. Velocity is slower in systole and falls more gradually during diastole, never reaching zero and always going forward through all cardiac cycles.

Flow parameters

All the information received from the Doppler spectrum is useful when deciding if the study vessel does or does not have flow pathology [7]. To do so, there are threshold values obtained from spectral analysis.

The main problem in accurately determining flow velocity is the Doppler angle. The estimation of it is very difficult when analyzing very tortuous vessels or very small ones. Although this absolute quantification of flow velocity is sometimes imprecise because of the failure to determine the Doppler angle, it is possible to calculate other flow parameters like the resistivity index (RI), pulsatility index (PI), and systolic-diastolic velocities ratio (S/D ratio).

As the flow parameters all come from the same Doppler spectrum, they are useful not because they give absolute quantification of flow, but because errors in Doppler angle can be normalized.

$$PI = (v_{max} - v_{min}) / v_{mean}$$

$$RI = (v_{max} - v_{min}) / v_{max}$$

The RI or Pourcelot index is given by subtracting the minimum (diastolic) velocity from the maximum (systolic) velocity, and dividing all by the maximum velocity. High values indicate a high-resistance blood flow vessel, which can be obtained by arteriole constriction or limited arterial distensibility [4, 7].

To evaluate the PI, mean velocity must be calculated first. Nowadays, ultrasounds scanners require the operator to mark just the maximum and minimum velocities on the spectrum, and after that the computer will calculate the mean velocity. The values are obtained by subtracting the minimum (diastolic) velocity from the maximum (systolic) velocity, and dividing the result by the mean velocity. Stenosis and high-resistance vessels will increase the PI values [7].

The systolic-diastolic ratio is a variation of the RI, dividing systolic velocity by diastolic velocity.

Further reading

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