Vascular Imaging of the Central Nervous System
Physical Principles, Clinical Applications, and Emerging Techniques
I dedicate this book to my husband Miguel, and to our children Carlota, Miguel, and Francisco, for their love and understanding. To my parents, Antonio and Maria Eugenia, who taught me the meaning of commitment and persistence and to my sister Rita, for always being present.

Joana N. Ramalho
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In recent years, numerous new developments have taken place in vascular imaging, which has evolved from invasive procedures to safer, faster, and more elegant, noninvasive, and precise techniques and methods. Rapid progress in cross-sectional modalities such as sonography, computed tomography, and especially magnetic resonance imaging has played a special role in these developments, which have translated into better patient management. Parallel to these important advances, several innovative and exciting techniques have been introduced to clinical practice. For example, it is now impossible to think about vascular imaging without considering cerebral perfusion-imaging techniques, vascular wall imaging (including atherosclerotic plaque assessment), and intravascular imaging. In just a few years, these methods have gained wide clinical application.

A huge range of vascular-imaging methods are now available for clinical practice, and radiologists are often faced with several choices. Unlike previous texts, which cover this topic according to disease processes, this book looks at vascular imaging from the standpoint of these various imaging techniques.

*Vascular Imaging of the Central Nervous System: Physical Principles, Clinical Applications, and Emerging Techniques* comprehensively covers all these new and fascinating diagnostic techniques and provides an up-to-date overview of their role in clinical vascular imaging. The overall aim of this book is to optimize the use of these techniques by practicing radiologists, facilitate vascular imaging, and thereby improve management of patients.

Each chapter includes first a description of the physical principles underlying each technique, as well as the pros and cons and associated possible artifacts and pitfalls. Then clinical applications of each method are presented, along with novel and possible future applications of each technique, wherever appropriate.

The increasing complexity of these imaging techniques requires higher-level knowledge about how to perform the studies effectively, how to recognize their limitations, and how to interpret them. We hope that this book provides some of that knowledge and proves truly useful to our colleagues.

Joana N. Ramalho and Mauricio Castillo
JR:
It was a pleasure to be involved in this project with Dr. Castillo, a dedicated teacher and a great mentor. His mentoring and inspiration have changed the course of my career, and I am forever indebted to him.

MC:
This book proves the ancient adage: “Give a man a fish; you have fed him for today. Teach a man to fish; and you have fed him for a lifetime.” The contents of the book were selected, edited, and approved by Dr. Joana Ramalho, a dedicated neuroradiologist with a bright future. I thank her profoundly for allowing me to participate in this exciting project.
Ultrasound Vascular Imaging (UVI)
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 ($\lambda$), 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 ($A_2 / A_1$). By halving the amplitude, the decibel’s level goes up by six (dB = 20log ($A_2 / A_1$)).

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 ($\rho$) 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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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.

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 ($\frac{\sin \theta_1}{\sin \theta_2} = \frac{c_1}{c_2}$).

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) ($I = \frac{W}{cm^2}$).

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: $\text{dB} = 10 \times \log_{10} I$.

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 $\alpha$ 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 1.3** 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.

**Figure 1.4** The incidence beam and the type of reflector.
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, heart-beats, 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):

\[ f_f = \frac{2 \times f_r \times v \times \cos\theta}{c} \]

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.
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...
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 beam, it will be seen in the negative part of the y-axis (below the zero baseline).

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
Chapter 1: Basic Principles of Ultrasound Sonography

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 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.

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 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.
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 artificiatively 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 × T × F_p × cosθ)).

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 (ρ) 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.

\[ Re = v \times d \times \rho / \eta \]
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-

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.
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.

\[ \text{PI} = \frac{(v_{\text{max}} - v_{\text{min}})}{v_{\text{mean}}} \]

\[ \text{RI} = \frac{(v_{\text{max}} - v_{\text{min}})}{v_{\text{max}}} \]

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


References

Introduction

Vascular ultrasound is a common, noninvasive method for the evaluation of extracranial and intracranial arteries, giving not only information about anatomy but also hemodynamic information about the vessels [1, 2]. Besides its noninvasiveness, vascular ultrasound has other advantages: it is a low-cost, readily available exam, which provides an important amount of information relatively quickly [2].

There are many clinical indications for noninvasive imaging of the cerebral arteries [2, 3]. In this chapter, we will discuss the main clinical entities for which cerebrovascular ultrasound is an important evaluation tool, particularly in the evaluation of extracranial and intracranial stenosis and occlusions, and cervical artery dissection.

Imaging protocols

Ultrasound examination allows visualization of anterior and posterior circulation from the common carotid arteries to the main basal cranial cerebral arteries, and the vertebral arteries.

Cervical duplex ultrasonography (CDU) uses a high-frequency transducer (typically 5–10 MHz) to examine the carotid arteries (including the common carotid artery, its bifurcation, and both internal and external carotid arteries) and the vertebral arteries [2, 4, 5]. CDU combines three scanning techniques: B-mode imaging, which allows evaluation of the arterial wall structures; color Doppler imaging that evaluates the vessel lumen; and the Doppler frequency spectrum, which allows determination of the velocity and direction of the flowing blood [2, 4, 5]. When beginning an examination of the cervical vessels, the patient should be supine with the head turned about 10–20° from the insonated side, and the examination should include B-mode, color mode, and Doppler spectrum analysis. Both axial and longitudinal planes should be obtained [5].

The common carotid arteries (CCAs) are relatively stable and symmetric vessels that can be examined over the 5 cm of length proximal to the carotid bifurcation. The vessel caliber is about 6–7 mm, and it can be distinguished from the jugular vein because the latter can be obstructed when compressed with the transducer. The CCA divides into the internal carotid artery (ICA) and external carotid artery (ECA). Some imaging findings can help distinguish the ICA from the ECA. Typically, the ECA is more superficial and has a smaller caliber than the ICA, and the ICA is usually posterolateral to the ECA. The ECA has extracranial branches, the first one being the superior thyroid artery, while the ICA has no extracranial branches. The spectral Doppler waveform is also different, with the ECA showing virtually no diastolic flow because it is a high-resistance vessel. Also, the ICA has a lower pulsability compared to the ECA. Of note is the irregular blood flow pattern on the carotid sinus, which causes a less laminar and lower velocity flow that is normal and is caused by the widening of the vessel in that location [5].

Examination of the vertebral artery (VA) should start on the V2 segment, which is easy to identify. The patient’s head should be in a straight position with the chin slightly elevated. As with the other vessels, B-mode, color mode, and Doppler spectrum evaluation should be performed. The best way to look for the vessel is by identifying the cervical transverse processes, which appear as hypoechoic structures, and the VA can be seen running perpendicular to them. The V0/V1 segment is more difficult to insonate due to its origin behind the clavicle. It should be located by following the V2 segment in the color mode. When studying the V3 segment, the patient’s head should be turned, with the transducer located below the mastoid, showing a comma-shaped structure in the color mode [5].
Noninvasive assessment of cerebral hemodynamics is made by transcranial Doppler sonography that uses a 2 MHz pulsed Doppler transducer, with penetration of the ultrasound beam through the bone.

This technique was first used to evaluate the cerebral vasospasm. Nowadays, it not only plays an important role in the detection and follow-up of vasospasm caused by subarachnoid hemorrhage, but it is also very important in the diagnosis of stenosis or occlusions of intracranial arteries, determination of brain death, and assessment of cerebral hemodynamics after trauma, stroke, or migraine [6].

Transcranial Doppler examination is performed with the patient in the supine position, looking forward, and with the examiner at the head of the bed. It is performed through four windows: temporal, orbital, suboccipital or transforaminal, and submandibular (Figure 2.1).

The **temporal window** is a region in the suprazygomatic portion of the temporal bone, where the skull is naturally thin. The dimension of each bone window is individually different and depends on several factors [7]:

- Thickness of the diploe: the diploe scatters the ultrasound beam, and its thickness increases with increasing width of the temporal bone.
- Gender: the temporal bone window is better in men than in women.
- Age: insonation quality decreases with age.

There are three temporal windows (Figure 2.2) [8]. The posterior window is immediately anterior to the external auditory canal, the middle window is 1.5 cm anterior to the posterior window, and the anterior window is 1.5 cm anterior to the middle window.

Through the temporal window, the velocity of the middle (Figure 2.3), anterior (Figure 2.4), and posterior (Figure 2.5) cerebral arteries and the terminal portion of the ICA can be measured. Anterior and posterior communicating arteries can also be detected when they exhibit accelerated blood flow velocity (collateral routes).

The **orbital window** is accessed by applying the transducer to the closed eyelid, using the lowest acoustic intensity. The ultrasound beam goes through the orbital window, passing by the superior orbital fissure, optic foramen, and orbital plates of the frontal bone, and it allows the study of the ophthalmic artery and the three segments of the carotid siphon.

The **transforaminal window**, or foramen magnum window, is accessed with the patient in sitting or supine position, with the patient’s head to the side and flexed...
forward. Through the foramen magnum, study of the intracranial portion of the VA (V4) and basilar artery can be made (Figure 2.6).

The **submandibular window** allows the study of the extradural segment of the ICA by placing the transducer below the angle of the mandible.

Transcranial Doppler examination depends on several factors [8]:
- Window
- Depth of sample volume (in millimeters)
- Direction of flow
- Angle of transducer in relation to the patient’s head

**Figure 2.3** Normal wave in middle cerebral artery (MCA), with flow toward the transducer, at a sample volume depth of 56 mm with a normal blood flow of 84 cm/s. ACA D, right anterior cerebral artery; Amostra, sample; Diast, diastolic velocity; MCA D, right middle cerebral artery; MCA E, left middle cerebral artery; Media, average; PCA P2 E, left posterior cerebral artery P2 segment; PI, pulsability index; Potencia, power; Profund, depth; RI, resistance index; Sist, systolic velocity; VA D, right vertebral artery.

**Figure 2.4** Normal wave in anterior cerebral artery (ACA); the flow is away from the transducer, at a sample volume depth of 66 mm with a normal blood flow of 50 cm/s. ACA D, right anterior cerebral artery; Amostra, sample; Diast, diastolic velocity; MCA D, right middle cerebral artery; MCA E, left middle cerebral artery; Media, average; PCA P2 E, left posterior cerebral artery P2 segment; PI, pulsability index; Potencia, power; Profund, depth; RI, resistance index; Sist, systolic velocity; VA D, right vertebral artery.
Figure 2.5  Cerebral posterior artery (P2), with flow away from the transducer at a sample volume of 66 mm with a normal blood flow of 38 cm/s. ACA D, right anterior cerebral artery; Amostra, sample; Diast, diastolic velocity; MCA D, right middle cerebral artery; MCA E, left middle cerebral artery; Media, average; PCA P2 E, left posterior cerebral artery P2 segment; PI, pulsability index; Potencia, power; Profund, depth; RI, resistance index; Sist, systolic velocity; VA D, right vertebral artery.

Figure 2.6  Normal wave of vertebral artery, with the transducer at suboccipital window and flow away from the transducer; sample volume is 70 mm. ACA D, right anterior cerebral artery; Amostra, sample; Diast, diastolic velocity; MCA D, right middle cerebral artery; MCA E, left middle cerebral artery; Media, average; PCA P2 E, left posterior cerebral artery P2 segment; PI, pulsability index; Potencia, power; Profund, depth; RI, resistance index; Sist, systolic velocity; VA D, right vertebral artery.
PART ONE Ultrasound Vascular Imaging (UVI)

• Relationship of the vessel to the middle cerebral artery (MCA), anterior cerebral artery (ACA), and terminal ICA
• Response to CCA compression (diminished, reverse, or absent flow indicates that the vessel is supplied by the CCA).

Transcranial Doppler examination allows a complete evaluation of the circle of Willis and vertebrobasilar system. It begins by insonating through the temporal window, and after insonating the transforaminal and submandibular windows.

In transcranial Doppler sonography, the diagnosis is based on the change of flow velocity, the absence of blood flow, or changes in pulsatility. Many factors affect the blood flow velocity and pulsatility of intracranial vessels, including age, hematocrit, blood viscosity, cardiac output, partial pressure of CO₂ and O₂, degree of brain activation, and cerebral metabolism [8]. These measurements can vary between individuals, in the same individual temporally, or due to the skills and experience of the examiner.

The evaluation of part or all of the intracranial vessels sometimes is not possible because of an absent temporal window (e.g., due to calvarial thickness) or due to unusual vessel position or tortuosity.

Assessment of cerebral hemodynamics also can be made by transcranial color duplex sonography that is performed with a phased array transducer, giving information about anatomic landmarks and special courses of the arteries that are important for vessel identification [9]. Color flow, power Doppler, and even three-dimensional imaging could be used complementarily with this duplex sonography examination.

The study is made through the same acoustic windows (i.e., transtemporal, orbital, and foramin). Another acoustic window, the frontal window, can be used to study the A2 segment of the ACA.

With transcranial B-mode imaging, the identification of the circle of Willis arteries is possible, giving important information about their course, tortuosity, and arterial branching. Color Doppler imaging shows the vessel, determine the direction of the blood flow and detects regions with disturbed flow. Power Doppler can help the visualization of the arteries and it is particularly useful in the morphologic characterization of aneurysms, intracranial focal stenosis or occlusions, and arteriovenous malformations.

Clinical applications

Extracranial vascular applications

Stenosis and occlusions

Cerebrovascular diseases are one of the main causes of mortality and morbidity in the Western world and developing countries. In the United States, stroke is the third leading cause of death. About 80–85% of strokes are ischemic [1, 4], and in 20–30% of these, atherosclerotic disease of the ICA is the cause [2, 4]. Stroke's overall social and economic burden requires an early diagnosis and a precise characterization, which are both necessary for its adequate prevention and treatment [2]. Other causes of stroke include cardioembolic disease (30%) and small-vessel disease (20%) [10], as well as less frequent causes such as nonatherosclerotic, non-hypertensive vascular diseases ( moyamoya disease, cranio-cervical arterial dissection, and primary and systemic vasculitis); hypercoagulable states; and hematologic disorders, among others [10].

About 80% of strokes are thromboembolic, and many times the embolic source is the plaque itself. Nonetheless, in about 45% of strokes in the territory of a symptomatic carotid stenosis, the cause is either cardioembolic or small-vessel disease [11, 12].

In the carotid system, the most frequent site of plaque formation is the distal primitive carotid artery and the first 2 cm of the ICA [11]. Other sites also commonly involved include the carotid siphon and the proximal anterior and middle cerebral arteries [11].

Cervical ultrasound is one of the methods that can be used in order to characterize a vessel morphologically and accurately measure the degree of stenosis [2, 13].

Morphologic assessment

When examining the arterial wall, the intima-media complex can be identified as a double-line structure. Its thickness, measured by B-mode echotomography, is considered a risk factor for stroke and myocardial infarction [4, 14] and, according to the GENIC (Etude du Profil Genetique de l’Infarctus Cerebral) study, is considered a predictive factor for lacunar stroke [4, 15] (Figure 2.7). The normal thickness depends on the age and risk factors of the patient, and values between 0.35 and 1.2 mm are usually considered normal [16].

Using B-mode echotomography, carotid plaques can also be identified, and their extent, texture, echogenicity, surface, total area, and volume can be characterized [2, 4]. A plaque is considered homogeneous when the echoes are uniform throughout the plaque and heterogeneous when there are variations in the echoes. These differences in echogenicity correlate with plaque composition (Figure 2.8). Typically, a plaque that is homogeneous is more cellular, while heterogeneity usually is caused by calcifications, cholesterol deposition, or hemorrhage. Anechoic or hypoechoic plaques are usually richer in lipids, and this has been suggested as a potential risk factor for stroke [2, 4, 15, 17]. Hyperechogenicity with distal shadowing has been associated with plaques containing calcifications. This acoustic shadow does not allow a correct visualization of the vessel beneath the plaque, and is considered a limitation of echotomography.