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PART I

I Principles and Instrumentation

CHAPTER 1

1

Principles of Vascular Laboratory Testing

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Vascular laboratory testing is performed in order to examine defined blood vessels and to characterize blood flow within these vessels. This requires an understanding of vascular anatomy, ultrasound imaging, physiologic factors governing blood flow, and display of Doppler analysis.

Ultrasound image generation

An ultrasound transducer generates sound waves in discrete pulses, which then travel through the soft tissue. A fraction of the sound waves is reflected back towards the transducer as it encounters a change in tissue acoustic properties. The position of the tissue interface encountered by the returning echo is determined by the duration of time between the transmission and return of the pulse. The strength of the returning signal is proportional to the density and size of the tissue causing the scatter. The strength and time of the returning echoes are used to construct the gray scale (B mode) image. Higher transducer frequency results in greater near-field contrast resolution and decreased depth of penetration as a result of attenuation. Specular reflection occurs when an interface is large and smooth with respect to the transmitted wavelength. Specular reflection is maximal when the ultrasound beam is perpendicular to the interface, and allows for the characterization of the vessel wall.

Images of the vessel wall may also be improved by the use of harmonics. The wide bandwidth of current vascular probes allows the analysis of returning harmonics (whole number multiples) of the fundamental frequency. Conventional ultrasound imaging sends out a fundamental sound wave of defined frequency and receives the same frequency range back in the returning echoes. The fundamental wave becomes distorted as the tissue compresses and expands in response to the sound wave. This distortion results in the generation of harmonics, additional frequencies that are multiples of the emitted frequency. The receiver can detect both the original frequency and the harmonics. When the receiver detects echoes only at the harmonic frequency, it can reduce artifacts associated with the fundamental frequency, such as speckle, reverberation, and side lobes[1] (Figure 1.1).

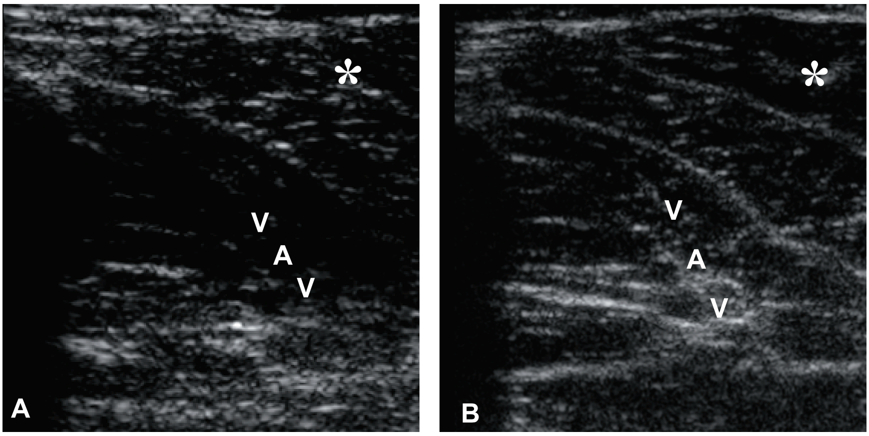

Figure 1.1 Gray scale (B mode) imaging. (A) Gray scale imaging of the posterior tibial vessels (VAV) in the calf utilizing the fundamental frequency. Near-field contrast and speckle (*) are both more evident with the fundamental frequency. (B) Imaging of the same posterior tibial vessels utilizing harmonics. The near-field resolution and speckle (*) are decreased. All interfaces appear more prominent. In this instance, harmonics allows improved detection of the posterior tibial artery and the paired posterior tibial veins.

Blood flow

Blood flow occurs when there is a difference in pressure between two points, and flow proceeds in the direction of the lower pressure[2]. Major regulation of flow occurs at the heart and at the resistance vessels. The cardiac pump cyclically restores high-pressure (high-energy) flow into the arterial system. Some energy is then lost by friction as blood flows through the arteries. The next major control point in regulating blood flow is the resistance vessels. Constriction of these vessels increases distal pressure and therefore decreases blood flow, while dilation decreases distal pressure and allows an increase in blood flow. The dynamic change possible at the level of the resistance vessels allows for a more precise regional control of blood flow to specific organs. Downstream from the resistance arterioles is the venous pool. Blood flow at this point in the system is at low pressure (energy). Venous flow is therefore more profoundly affected by the contribution of hydrostatic and intrathoracic pressures. The downstream veins are progressively larger in diameter and easily distensible; therefore, there is little resistance to forward flow. The lowest pressure (energy) in the circulatory system is in the right atrium, which normally is close to atmospheric pressure.

Laminar and nonlaminar flow

In a straight vessel with uniform diameter, blood moves forward in concentric circles with the middle circle having the highest velocity and the outermost circle having the lowest velocity. The velocity profile across the vessel is a

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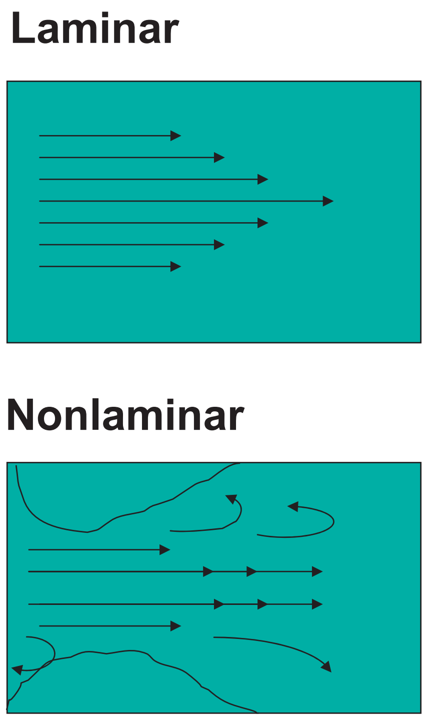

Figure 1.2 Laminar flow is characterized by the highest velocity in the center stream (long arrow) and the lowest velocity at the wall (short arrow), with a parabolic decline in velocity from mid-stream to the wall edge. Nonlaminar flow includes chaotic variations in both velocity and flow direction, as demonstrated by the arrows. It results from changes in the tube diameter.

parabola, as the rate of change in velocity is greatest against the wall and least in the center. The flow velocity changes throughout the cardiac cycle, with the highest velocity during systole and the lowest velocity during diastole. There are normal and abnormal deviations from laminar flow in the circulation. The lines or circles of flow are disturbed when the vessel diameter changes and at bends or branches (Figure 1.2). Disturbances in laminar flow are referred to as turbulence[3] and are described by the dimensionless Reynolds number. Turbulence is detected on physical examination by the presence of a bruit or a thrill. Pressure decline along the length of the vessel is greater when there is turbulent flow rather than laminar flow.

Poiseuille's law

The mean velocity of laminar flow in a cylinder is directly proportional to the energy (pressure) difference between the two ends of the cylinder and to the square of the radius. The velocity is inversely proportional to the length of the tube and the blood viscosity. The most important feature of blood flow is the volume of the flow delivered, and the volume is proportional to the fourth power of the blood vessel radius. A decrease of 50% in the radius results in a 95% decrease in flow. As the length of the vessels and the viscosity of blood in the cardiovascular system do not change, it is alterations in radius and decreases in inflow pressure that limit the volume of blood flow. The resistance is the quantification of the difficulty in forcing blood through the vessels. The resistance is also derived from Poiseuille's law when the pressure differences and blood flow can be measured [4]:

$$\text{Volume of flow} = [\pi(P_1 - P_2)r^4]/8L\eta$$
$$\text{Resistance} = 8L\eta/\pi r^4 = (P_1 - P_2)/Q$$

where P is the pressure, r is the radius, L is the length, η is the viscosity, and Q is the volume flow.

Pulsatile pressure and flow

One stroke volume of blood is ejected into the arterial system with each heartbeat. The waveform of the ejected blood changes as it travels through the arterial system. During late systole, when the ejection volume decreases, the outflow volume through the peripheral resistance vessels exceeds the volume being ejected by the heart, and the pressure begins to decline. The energy stored by arterial distension also decreases throughout the cardiac cycle. The speed at which the wave travels is referred to as the pulse wave velocity. The speed of the wave is dependent on the vessel distensibility and is independent of the volume of blood flow. The amplitude of the pressure wave and the systolic pressure increase as the wave travels distally in the aorta as a result of reflected waves. Reflected waves occur with branching, changes in vessel dimensions, and changes in vascular stiffness. The reflected waves in the periphery are enhanced because of the high resistance typically encountered, and result in high thigh pressures normally greater than the ankle or brachial pressures. This is relevant to the correct interpretation of pressure measurements in peripheral arterial evaluation. As the ankle systolic pressure is normally higher than the brachial pressure, the normal ankle : brachial ratio (index) is greater than 1.0. The waveform changes when the artery is flowing into an organ with dilated resistance vessels. Forward flow is detected throughout diastole in this setting of low resistance (Figure 1.3).

Arterial stenosis

Arterial stenosis, or narrowing, can result in decreased pressure and flow downstream. A diameter decrease of 50% results in a cross-sectional area (CSA) decrease of 75%. Whether or not this results in hemodynamic change is influenced by the presence of turbulence, the ratio of stenosis CSA to more proximal CSA, the rate of flow, and the peripheral resistance. The concept of "critical stenosis" is a simplification of the complex interplay of factors. For example, serial stenoses have a greater impact on distal pressure and flow than a single stenosis. A decrease in pressure distal to the stenosis is most illustrative of the hemodynamic impact of the stenosis. Newer techniques, such as the evaluation of the

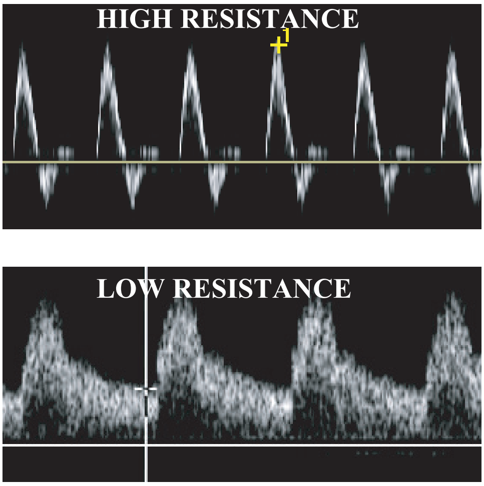

Figure 1.3 A high-resistance arterial waveform is characterized by a systolic peak and the absence of flow during some portion of diastole. A lower resistance arterial waveform is characterized by peak systolic velocity followed by continued flow during diastole until the next systolic peak.

transit of microbubbles across an organ or tissue, are also being used to determine the impact of hemodynamic alterations[5].

Venous waveforms

Phasic changes in the low-pressure venous system are a result of changes in the intrathoracic pressure with respiration and changes in right atrial pressures. The changes in flow that accompany respiration are opposite in the upper extremities when compared with the lower extremities. In the upper limbs, flow increases during inspiration, when the right atrial pressure is lowest. In the lower limbs, abdominal pressure increases during inspiration and the lower limb venous flow decreases. The amount of venous flow is also governed by the flow into the venous system. The resistance arterioles are dilated when limb blood flow increases, and venous flow therefore increases. In states of increased venous blood flow, the phasic changes may be evident (Figure 1.4). Flow and pressure changes during the cardiac cycle are most evident in the central veins[6] and are represented by three positive pressure waves. These are referred to as the a, c, and v waves and occur coincident with atrial contraction, atrial ventricular valve closure, and atrial filling, respectively. Cycle variation in the peripheral venous waveform is most evident in right heart failure when the venous pressure is increased. Cyclic variation is also apparent in healthy "well-hydrated" individuals when the venous system is fully distended.

Hydrostatic pressure and posture

Hydrostatic pressure arises from the gravitational potential energy of the blood.

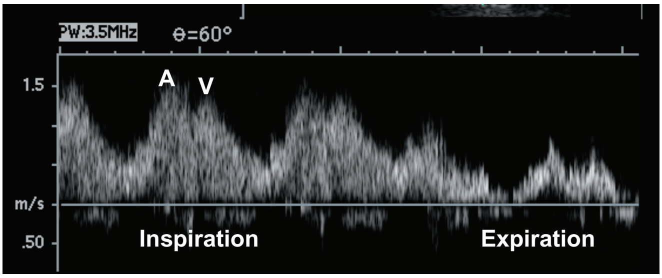

Figure 1.4 Spectral display of internal jugular venous flow. The a and v waves are evident. The velocity varies in response to the respiratory cycle, increasing during inspiration.

It occurs within the body when one part of the body is at a different elevation from another. In the upright position, the hydrostatic pressure is greatest in the lower extremities, where the additional gravitational force is the greatest. This results in venous distension and a further increase in the amount of blood pooled in the lower extremities. Contraction of the surrounding muscles increases pressure within the veins and propels blood towards the heart (as long as the one-way venous valves are competent to prevent flow away from the heart). The systolic and diastolic pressures at the ankle increase by up to 100 mmHg in the standing position, compared with the supine position, as a result of the contribution of hydrostatic pressure.

Display of Doppler information

Most images of blood flow rely on the detection and processing of Doppler shift frequencies[7]. The frequency of echoes returning after striking a moving object will be different from the frequency emitted by the transducer. The shift in frequency will be positive or negative depending on the direction of flow. The difference between the transmitted and received frequencies is affected by the speed of sound c, the flow velocity v, and the angle between the insonation beam and the direction of blood flow θ (Figure 1.5). These relationships are described by the Doppler equation:

$$f_{\text{shift}} = f_{\text{received}} - f_{\text{transmitted}} = (2f_{\text{transmitted}}\, v \cos\theta)/c$$

The Doppler angle θ therefore has a strong influence on the detected Doppler shift frequency for a given reflector velocity. In practice, the post-processing on current ultrasound machines will use the measured Doppler shift frequency to calculate the blood flow velocity:

$$v = c(f_{\text{received}} - f_{\text{transmitted}})/2f_{\text{transmitted}} \cos\theta$$

Spectral Doppler systems produce sound or graphic displays to represent the detected Doppler shift (Figure 1.6). The graphic displays typically have velocity

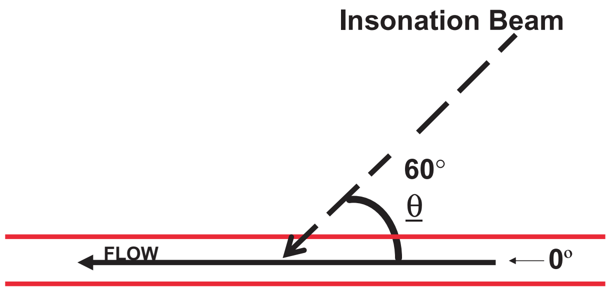

Figure 1.5 The Doppler angle refers to the angle between the insonation beam and the flow. At 0°, the insonation beam is parallel to the flow. This insonation angle is not generally achievable in vascular imaging. The angle of 60° is easily obtained in most blood vessels.

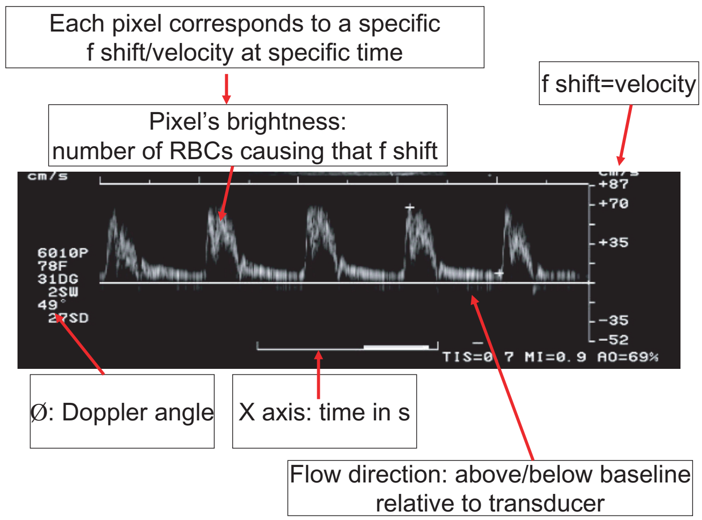

Figure 1.6 Spectral display of pulsed wave Doppler interrogation. The amount of information contained in the display is evident from a review of the many components indicated in this figure.

on the *y* axis and time on the *x* axis. Color Doppler displays the detected Doppler shift by colors indicating the velocity and direction of flow. Phasic continuous wave (CW) Doppler allows the detection of the magnitude of the Doppler shift, but not the direction. Reflectors and scatterers anywhere in the CW insonation beam contribute to the instantaneous Doppler signal. Therefore, with CW, Doppler shifts from multiple vessels can be sampled simultaneously. Pulsed wave (PW) Doppler samples and color encodes only the frequency shifts from a defined volume. The Doppler signal produced is generated from the frequency shifts (from moving targets) from one pulse echo sequence to the next. The pulse repetition frequency (PRF) must be high enough so that important details of the

Doppler signal are not lost between transmitted pulses. Aliasing occurs when Doppler shifts take place during transmitting intervals, but not during receiving intervals. The limit of the velocity display is determined by the PRF. No velocity above a specific Doppler threshold is displayed when the Doppler shift threshold is greater than one-half of the PRF. This Doppler shift threshold is referred to as the Nyquist limit. Duplex instruments allow the display of the static B mode (gray scale) image simultaneously with spectral display of the PW Doppler signal.

Doppler angle of 60°

Velocity recordings are ideally obtained with an angle of 60° between the Doppler insonation beam and the vessel wall. This is achievable and reproducible in most vascular beds (Figure 1.5). The maximal frequency shift is detected at a Doppler angle of 0°, as is used in echocardiography. This Doppler angle cannot reliably be used in vascular imaging, as the vessels are parallel rather than perpendicular to the surface. The sample volume cursor is placed parallel to the wall, and a Doppler angle from 30° to 60° between the wall and the insonation beam (or flow jet) is utilized. At angles of less than 60°, small errors in setting the angle can result in up to a 10% error in velocity estimation, while at angles of greater than 60°, small errors in setting the angle can cause up to a 25% error in the velocity estimation (Figure 1.7). The decreasing value of the cosine at the higher Doppler angles has an increasing influence on the detected Doppler shift frequency, and therefore on the velocity estimation.

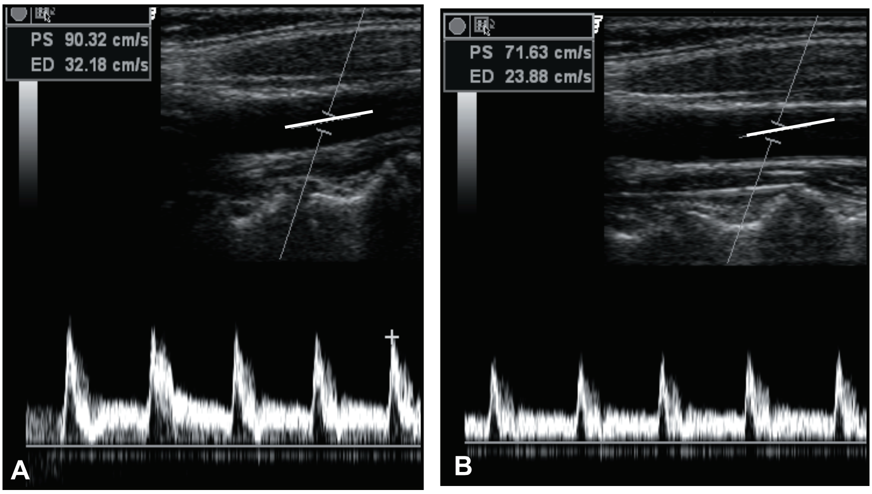

Figure 1.7 Impact of correct Doppler angle. (A) The angle between the flow and the insonation beam is 60° and the peak systolic velocity is 90.32 cm s^{-1} (top left). (B) The angle is 60° but the sample cursor is not aligned with either the flow or the vessel wall. The resulting velocity is 71.63 cm s^{-1}, 20% less than the velocity determination with the correct alignment.

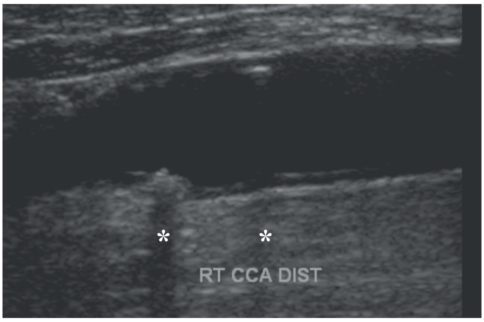

Figure 1.8 Shadowing artifact. The asterisks indicate the region of a shadowing artifact deep in an area of vascular calcification. The decreased echoes deep in the area of calcification are more prominent for the dense calcification on the left.

Artifacts

Errors in acoustic presentation occur because of both anatomic and technical factors. Shadowing occurs distal to dense objects (Figure 1.8). The image is less bright in the region being interrogated by fewer sound waves. In contrast, enhancement of the gray scale occurs distal to echo-free or liquid-filled structures. Multiple reflections from an interface (reverberation) can result in the addition of structures to the image. Refraction of the ultrasound beam can lead to improper placement of the structure on an image, and shadowing at the edge of large structures. Mirror image artifacts are produced when an object is proximal to a highly reflective surface, such as the pleura. Interaction with the reflector alters the timing of the returning echoes. The resulting image display includes a false echo equidistant from the reflector.

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