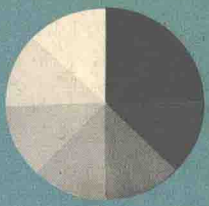
CLINICS IN DIAGNOSTIC ULTRASOUND 13



Vascular and Doppler Ultrasound

Edited by

C. Carl Jaffe, M.D.

CHURCHILL LIVINGSTONE

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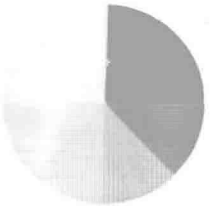

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VOLUME 13

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Preface

In the last several years there has been increasing interest in expanding the horizon of ultrasonic imaging to include extraction of the Doppler signal from structures that contain flowing substances, particularly blood. Although Doppler has a long history of use in a nonimaging mode by obstetricians for the detection of fetal heartbeat and in peripheral vascular examinations for the presence of flow in the ankle vessels, new commercial instrumentation extends the ability to perform Duplex Doppler examinations, which allow the Doppler sample to be directed at specific vessels. Not only is it potentially possible to determine the direction of flow but there is also an opportunity, under special circumstances, to allow a quantitative estimate of its velocity. This volume contains a variety of applications of both continuous- and pulsed-wave Doppler instrumentation as they are currently used by a variety of clinicians. The methods are still in their infancy and much wider clinical experience will be required before their proper role in diagnosis can be adequately evaluated.

The early investigations reported in these chapters are very encouraging and seem certain to secure Doppler a uniquely useful place among diagnostic methods.

C. Carl Jaffe, M.D.

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1 Doppler Applications and Limits of the Method

C. CARL JAFFE

Although the potential for recording Doppler signals from moving tissues has existed for over 20 years, medical applications other than for obstetrical heartbeat and peripheral vascular studies have been more modest than comparable progress in B-scan ultrasound. Compared to the latter, the Doppler signal is less intuitive, since it lacks static anatomic correspondence. This stems from the fact that the Doppler signal arises only from usually invisible flowing or moving substances within vessels. The signal is complex because it is inherently four dimensional and is composed of: (1) the Doppler shift frequency related to the velocity of the sampled structure; (2) the amplitude of that frequency; (3) a spatial distribution, since abnormal flow is usually nonuniform across the vessel; and (4) rapid temporal variation, since flow in most vessels is pulsatile. Attempting to comprehend all these variables at once is difficult and simplifications are necessary. The most easily understood method is a graphic waveform presentation[1] (Fig. 1.1), but the spatial position of the pulsed Doppler sample must be specified by viewing a second display composed of the two-dimensional B-scan image with the position and angle of the sample volume superimposed over the static image (Fig. 1.2). As an alternative to the waveform display for certain types of medical problems, such as carotid stenosis, the Doppler signal can be simplified by presenting the Doppler frequency–spatial position plot as a two-dimensional image[2] (Fig. 1.3). Qualitative display of the velocity pattern in specific vessels can be readily understood where abnormality is evidenced by higher peak velocity (brighter intensity) and narrowed vessels.

The common thread that connects most medical Doppler is that it is a technology capable of profiling the pattern of flow within specifically identifiable vessels. Therefore, with the exception of some peripheral vascular applications described in other chapters in this volume, it is usually necessary to develop instrumentation that permits selection of the Doppler sample from individual vessels. As a result, continuous-wave Doppler, which registers the Doppler

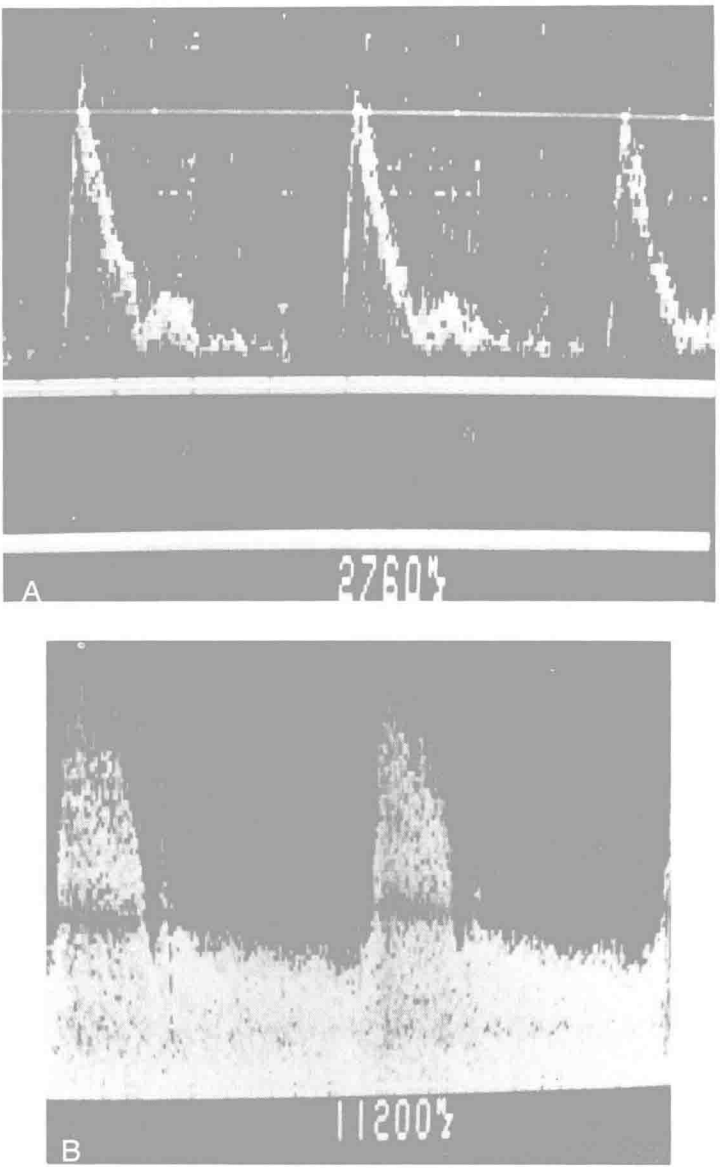


FIGURE 1.1. (A) Normal Doppler velocity pattern from the aorta. This temporal waveform display is the Fourier transform of the signal and shows that the peak systolic velocity, which occurs early in systole, is 2760 Hz. There is relatively little diastolic velocity. The waveform is clean and shows very little spread of frequencies indicating lack of substantial turbulence. The amplitude of the frequencies are represented as gray-scale intensity. (B) Fourier-transformed Doppler signal from a stenotic vessel shows that the systolic portion is composed of a spread of frequencies, indicating turbulence. The peak velocity is increased to greater than 11,000 Hz because of the jet effect at the stenotic orifice and there is a substantial amount of residual velocity during diastole because of the stenosis.

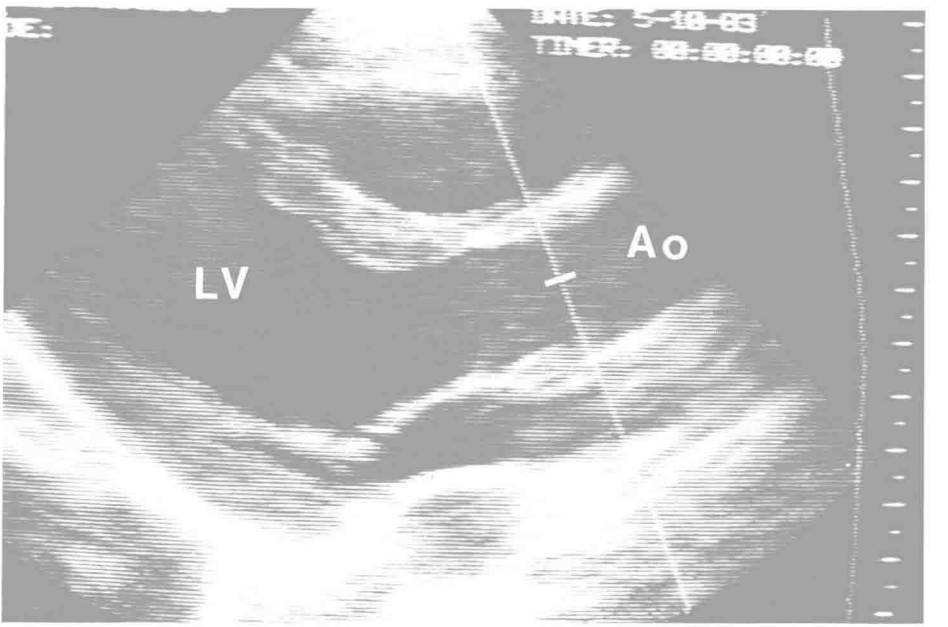


FIGURE 1.2. Duplex pulsed Doppler systems combine a B-scan image with a cursor, permitting directional and depth indicators of the position of the pulse Doppler sample. The figure shows a long axis view of the heart, displaying the left ventricle (LV) and aorta (AO). The Doppler sample volume can be selected along a directable cursor line at an adjustable depth (white bar). Here the sample is taken from the ascending aorta just above the aortic valve.

shift from all vessels along the beam axis indiscriminately, is inherently less attractive than duplex-pulsed echo techniques developed in recent years.[1,3] The duplex method involves using a two-dimensional static or real-time B-scan image to select a transducer position and depth from which pulsed samples of the reflected acoustic wave are examined for Doppler shifts. It should be apparent that a variety of compromises and simplifications are required with such a complex signal.

The most important compromise made in the development of duplex systems is that there is an inherent tension between those operational aspects that are most favorable for two-dimensional imaging and those required for optimizing the Doppler shift measurement. Conventional two-dimensional ultrasonic images of a vessel are best achieved when the vessel lies nearly perpendicular to the ultrasound beam axis, since then the vessel walls reflect the beam in a specular manner. If the pulsed Doppler signal is acquired from the same transducer position used for imaging, the sample creates a fairly large angle, θ, between the acoustic beam direction and the axis of flow (Fig. 1.4). As the angle θ approaches 90 degrees, the cosine, and therefore the Doppler shift frequency, reduces toward zero, as is evident from the standard equation:

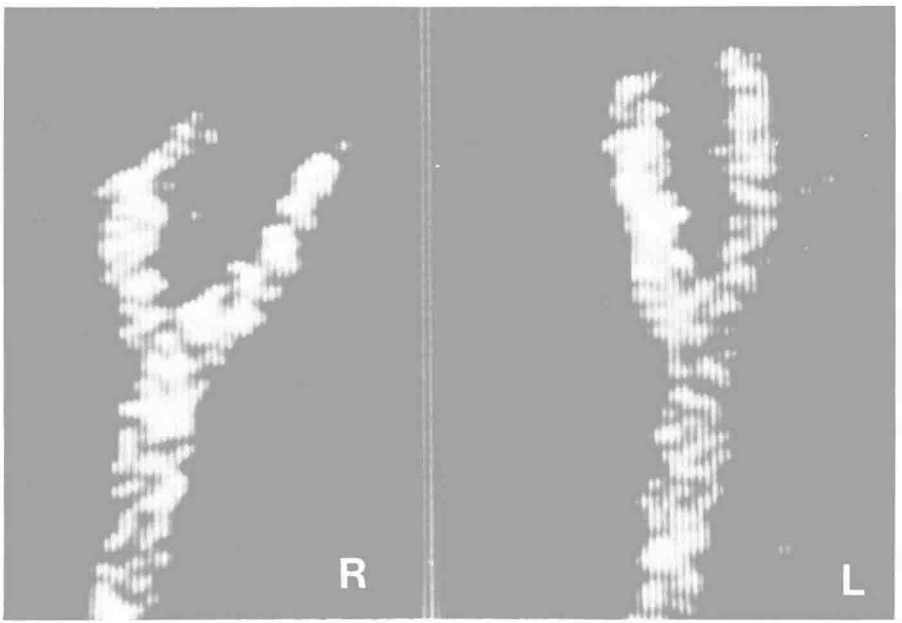
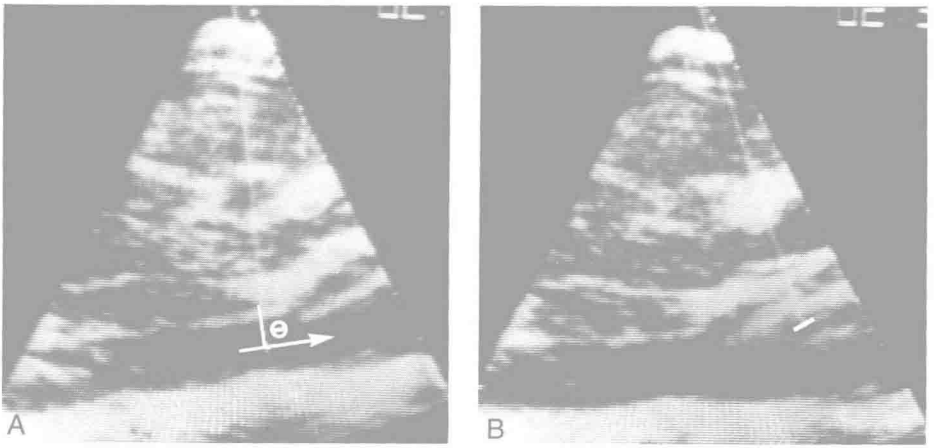


FIGURE 1.3. For simple nonquanitative applications, the Doppler signal can be presented as a two-dimensional display of the peak velocities. Here the right (R) and left (L) carotid bifurcations are shown with the peak Doppler shift velocity at specific depths displayed as intensity-modulated values. Many commercial systems use color encoding, in which high velocities are displayed as a different color. Diagnosis is made by noting not only vessel narrowing or vessel absence but also higher peak velocities encoded as a brighter intensity, which will occur in the region of a narrowed orifice.

$$\Delta f = \frac{2f_0 V \cos\theta}{c},$$

where Δf is the frequency change, f_0 is the original frequency, v is the velocity of flow, c is the speed of sound in tissue, and θ is the angle between the blood flow and the Doppler sample line.

Even when one is fortuitously able to image a vessel nearly along its axis, thus minimizing θ and maximizing the measurable Doppler shift, other difficulties arise. Because the Doppler shift is acquired from sequential short pulses a few cycles in length, accurate velocity information cannot be obtained if the repetition rate of these pulses is low. Therefore it is recognized that at deeper sample depths, the travel time required for the round trip limits the pulse repetition rate and thus limits the maximum Doppler shift measurable. As an example, Table 1.1 indicates the maximum Doppler shift frequency in kHz with comparable maximum velocities for specific sample depths. It is clear that the maximum detectable velocity of blood flow sampled with a 3-MHz signal along its axis ($\theta = 0$) and a depth of 13 cm would be approximately



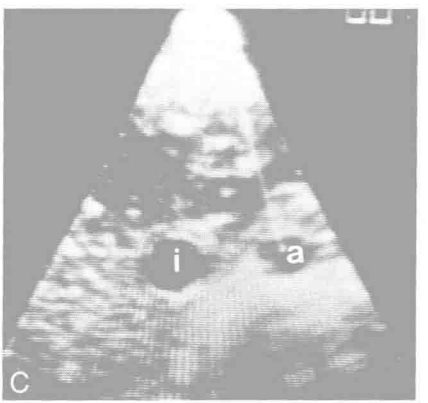


FIGURE 1.4. (A) Duplex Doppler sample extracted from sagittal view of the descending aorta. The direction of flow (arrow) intersects the direction of the sample at a large angle θ. The measurable vector component of the velocity therefore will give rise to a lower Doppler shift frequency than if the sample had intersected the vessel more along its axis. (B) Samples taken from the inferior mesenteric artery (bar) intersect that vessel at a smaller angle and therefore the detected Doppler frequency shift could be higher. (C) Transverse view of the inferior vena cava (I) with the Doppler sample being extracted for the descending aorta (A). Since both vessels are seen in cross section, it would not be possible to accurately estimate the angle of incidence of the sample relative to the long axis of the vessel. Signals extracted in this way would therefore be difficult to quantitate accurately.

70 cm/sec. A 5-MHz pulse at the same depth could measure maximum velocities only 60 percent of that value. Since many important arterial vessels contain blood velocities above that value, it is evident that it may be necessary to sample the vessel from a less acute angle, thus increasing θ to decrease the value of the cosine. Another simplification is imposed by the physical design of many systems, since it is often necessary to use the same transducer for acquiring the Doppler signal as for creating the B-scan image. Under these circumstances, the image is acquired, often with a real-time device with the B-scan image frozen in the display memory. A cursor line directs the transducer to sample the vessel at a specific angle and depth. It should be noted that in this method the Doppler information is being acquired at a spatial position that was frozen several seconds, and perhaps even minutes, earlier. Unless a

TABLE 1.1 Detectable velocity versus sample depth.

Maximum Δf KHz	Maximum sample depth (cm)	Approximate maximum detectable velocity (cm/sec at 3 MHz)
6.4	5	175
3.8	9	90
2.7	13	70

separate Doppler transducer or phased array technology is employed, it is impossible to both create a two-dimensional image and acquire a Doppler sample from a single acoustic line simultaneously. This obviously creates problems when vascular motion is vigorous. One of the most obvious examples of this difficulty can be demonstrated in cardiac applications, where the most favorable axial approach to the ascending aorta occurs from the apical four-chamber view (Fig. 1.5). This view places the sample just above the aortic valve. There is a double difficulty with this position since the sample often lies at depths greater than 12 cm and measurement of maximum blood flow velocity may be limited due to the lower pulse repetition rate required for sampling from a greater depth. Moreover, when sample placement uses the frozen B-scan image, inherent motion of the vessel during the cardiac cycle is often hidden. Positional drift of the sample volume between systole and diastole may be obscured. Possible corrective measures suggest themselves but each contains other limitations. The first is to use an "outboard" transducer, which allows the Doppler signal to be pulsed and sampled independent of the imaging transducer. Since the outboard Doppler transducer does not lie fully within the acoustic imaging plane, it transects that plane only at a specific depth. Nonetheless this approach has been implemented by some manufacturers. Although this solution is attractive, since the operator can keep the vessel continuously in view, it does not prevent the sample position from being shifted within the vessel lumen as it moves during the cardiac cycle. Another proposed solution is to use a time-varying function to modulate, and therefore track, the pulsed sample volume.[4] Alternative techniques use steered beam phased array imaging and have been implemented mostly on cardiac instruments. The steered beam phased array devices have the advantage of electively reducing the number of sampled lines for image creation and using the remaining time to direct the beam along a selected angular direction to acquire the Doppler sample. Although the two-dimensional image is made up of fewer individual lines, the image is not seriously degraded and allows dynamic visualization of the position of the Doppler sample volume. Though attractive, this solution still shares the common problem of an imposed maximum velocity limitation set by the lower pulse repetition rate required to sample a position of greater depth.

Concerning the sample volume, the user of Doppler systems should be aware