Physical and Technical Principles of Doppler Sonography

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Contents

Abstract

The frequency shift postulated in 1842 by Christian Doppler as a consequence of a relative movement between wave transmitters and receivers was only utilized for the first time to capture intra-corporeal movements in the middle of the 20th century. During these initial experiments, Satomura first used an ultrasound beam focused on the heart to measure the contractile movements of the myocardium and later demonstrated that this method can also be used to detect blood flow.

When exposed to ultrasound, the frequency reflected by the corpuscular blood components changes as the reflectors approach the probe or move away from it. Since this frequency shift is proportionate to the flow velocity, by comparing the transmission and reception frequency, the velocity can be determined and displayed in the form of a spectral curve.

In CW (Continuous Wade) mode, all flows within the beam are captured, while PW (Pulsed Wave) mode allows the selective detection of flows along a narrowly circumscribed space known as the sample volume. To be able to visualize this sample volume, PW Dopplers are generally combined with imaging ultrasound equipment to create "duplex systems" and the measuring window displayed in the form of a cursor in the B image.

The color-coded Doppler creates a number of tiny measuring locations and selectively displays the documented flows in the form of corresponding pixels on a screen. The color

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(usually red and blue) allows conclusions to be drawn about the direction of flow, while the flow velocity is expressed by the brightness of the corresponding pixels.

Echo signals that do not have their frequency shifted are evaluated as reflections of static structures and are displayed in the form of gray values. As a result, the color-coded (Color Flow) mode corresponds to a B-image with the location-correct display of flow information.

The Doppler ultrasound capture of flows is to a large degree dependent on the angle between the ultrasound beam and the flow axis, can sometimes fail when capturing very high velocities and demonstrates only limited spatial resolution. Additional alternative methods have also recently become available that are based on subtraction methods and which go some way towards overcoming the described shortcomings.

1.1 Doppler Effect

When Christian Doppler first presented his famous work *Über das farbige Licht der Doppelsterne und einiger anderer Gestirne des Himmels* (*On the coloured light of binary stars and some other stars of the heavens*) (Doppler 1842), the sceptical experts of the time were confronted with an unexpected thesis: The frequency which arrives at a receiver does not necessarily have to be the same as the one actually produced by the source. Doppler's hypothesis was a first attempt to explain the cause of the puzzling colour fluctuations of heavenly bodies circling one another. Inspired by references to the wave-form nature of light published previously, Doppler recognised the relative nature of the wavelengths detected and postulated that movements of either the wave generator or the observer changing the distance between them produce shifts in the frequency. Although Doppler's explanation for the cause of the colour fluctuations later turned out to be a misinterpretation, the basic idea of his thesis for the connection between movement and frequency detected proved to be an ingenious discovery. However, Doppler was not able to prove that his hypothesis was correct at first. It was someone else's work (Buys-Ballot 1845; Mach 1873), which showed that the equation suggested by Doppler for calculating the frequency shift was actually correct. In this connection, the proportional relationship between the velocity and resulting shift in frequency proved to be extremely important – it enabled the velocity to be determined accurately and simply by comparing the source frequency with the frequency received.

Technical developments from the middle of the twentieth century showed that it was also possible to determine the speed of objects which did not actively generate a frequency. Accurate measurements could also be made by introducing energy from outside and using the reflected frequency.

Even now, the method of choice for determining the speed of an object, gas or other fluid in scientific or applied technology applications is still the Doppler technique for detecting movements. The Doppler effect is also still used to quantify blood flow velocities. Although new techniques (e.g. B-flow) which can detect and image the corpuscular blood components directly without using the Doppler effect produce a significantly more highly resolved and highly realistic image of the spatial blood vessel path and blood flow, they cannot be used to quantify the speed.

As everybody knows, it is the Doppler effect on sound waves which is used to detect arterial or venal flows because they show particularly large frequency shifts due to their relatively low propagation rate, are regarded as non-invasive according to current understanding and are reflected by the corpuscular blood components in the required manner.

The three image sequences in Fig. 1.1 show the frequency shift produced by the Doppler effect. The left sequence is an example of sound radiation at 1,000 Hz with the sound generator in a fixed position. During each millisecond, the oscillator exerts a short positive pressure pulse on the environment followed by a negative pressure pulse of equal duration caused by the oscillator snapping back. These pressure changes are also called acoustic waves and propagate into the surrounding medium at a speed determined by the nature of the medium. The length of the acoustic wave is determined by the frequency of oscilla-

tion and the propagation rate in the medium – the wavelength of sound at 1,000 Hz described here, for example, is 34 cm long in air.

The middle sequence of Fig. 1.1 shows a sound generator operating at the same frequency but moving to the right at a uniform speed of 100 m/s. At the beginning of the sequence, the sound generator is on the left and produces the sound pressure zone designated '1'. After 1 ms, this part of the sound wave is propagated 34 cm to the right due to the speed determined by the medium. However, the sound generator has also moved by 10 cm and is now already starting to emit the next pressure wave. In comparison to the stationary sound source, this second sound pressure zone has been displaced by 10 cm to the right so the spatial distance between the two pressure zones which have been produced one after the other (i.e. the wavelength) is of course shortened.

The consequence of this reduced distance is that the pressure maxima arrive at the receiver with shorter time intervals so the receiver receives more pressure zones per second than would have been the case had the sound source remained stationary. The perceived frequency is therefore higher than the frequency actually produced.

In the example described here, the sound frequency received would be 1,416 Hz instead of 1,000 Hz which was the frequency at which the sound was generated. As Doppler correctly postulated, the displacement speed of the sound source is proportional to the change in frequency.

If, on the other hand, the sound source moves away from a receiver, the distance between two sound pressure zones will increase (Fig. 1.1 right). In this example also, the sound pressure zone designated '1' is emitted first and has moved about 34 cm to the right 1 ms later, due to the speed of sound in air. At the same time, the sound source has moved around 10 cm to the left and started to emit the next pressure zone from this point. The distance of 34 cm between the two pressure waves has therefore lengthened to 44 cm. A receiver would therefore receive the pressure waves with longer time intervals and would therefore register a lower frequency of just 773 Hz.

Changing the location of a sound source, however, is not the only way a frequency shift may be

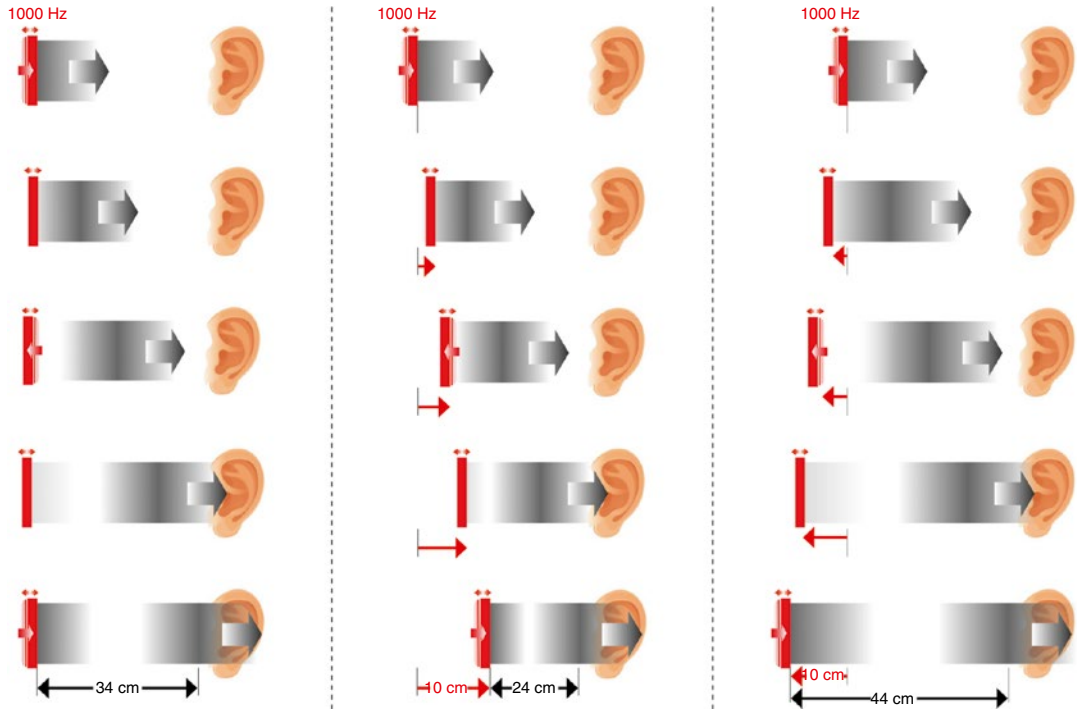

Fig. 1.1 (*Left*) Generation and propagation of sound waves. (*Centre*) Shortening of sound waves caused by an approaching acoustic source. (*Right*) Prolongation of sound waves when the acoustic source is moving away

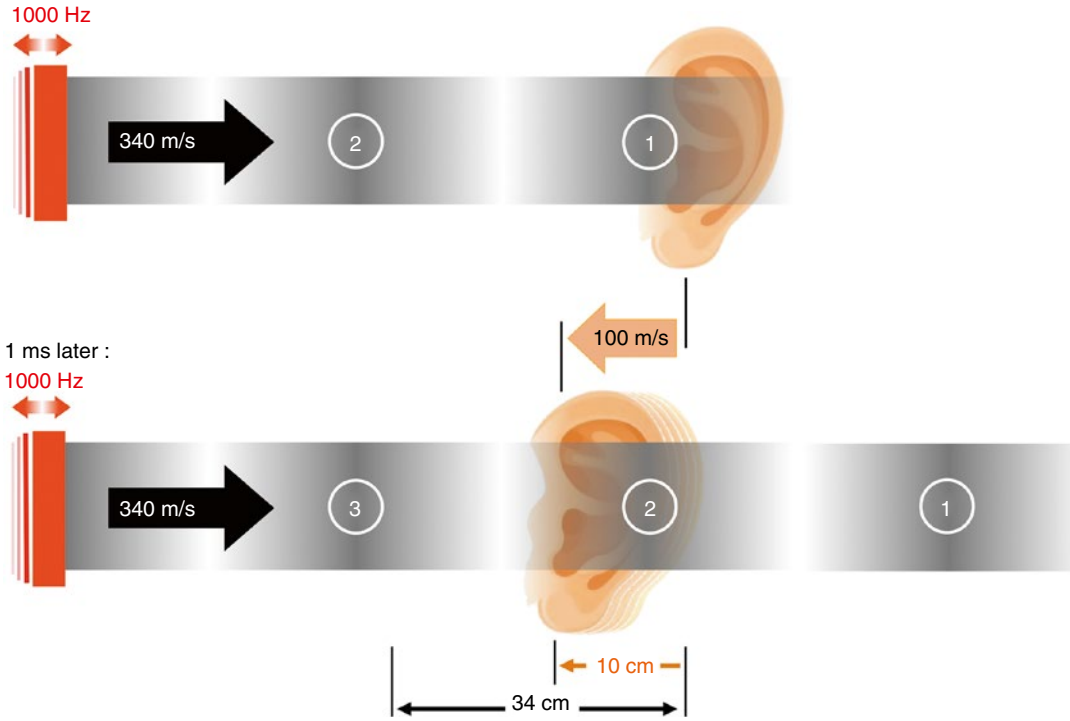

Fig. 1.2 Increase of observed frequency when the receiver is moving towards

produced – a corresponding effect is also experienced by moving the receiver.

This phenomenon is illustrated in Fig. 1.2.

The receiver in the figure moves at a speed of 100 m/s towards the approaching sound waves. The top of the figure shows that the pressure maximum designated '1' arrives at the observer. The high pressure zone designated '2' follows at a distance of 34 cm (in accordance with a sonic frequency of 1,000 Hz) and would normally reach the observer 1 ms later. Since the latter has moved closer to the sound generator by around 10 cm, the sound wave '2' arrives correspondingly earlier – i.e. after less than a millisecond. The shortening of this interval corresponds to an increase in the perceived frequency from 1,000 to 1,294 Hz.

On the other hand, if the observer moves away from the transmitter, the number of sound waves detected per second decreases. This is clearly shown in Fig. 1.3. At the top of the figure, the observer hurrying away receives the sound wave designated '1' at a frequency of 1,000 Hz. The wave designated '2' generated after a millisecond has to travel a longer distance than sound wave '1', since the observer has now moved 10 cm further away from the sound source. In this case, the interval between two reception cycles thus increased corresponds to a reduction in frequency from 1,000 to 706 Hz.

If the departure velocity were to increase to the speed of sound, the observer would find himself moving forwards with the sound wave inside a constant pressure zone. In this extreme case, periodic pressure changes would cease to arrive at the receiver and the frequency detected would be zero.

1.1.1 Summary

- Transmitter approaches receiver = increase in perceived frequency
- Transmitter moves away from receiver = decrease in perceived frequency
- Receiver approaches transmitter = increase in perceived frequency

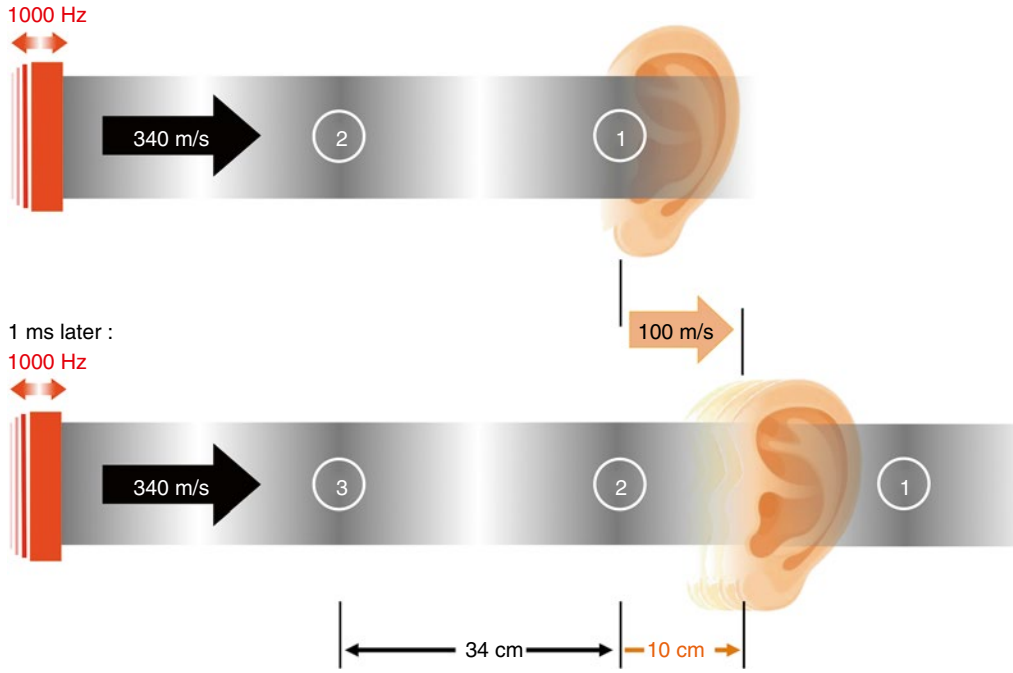

Fig. 1.3 Decrease of observed frequency when the receiver is moving away

- Receiver moves away from transmitter = decrease in perceived frequency

Since the situation is different in each case, Christian Doppler derived a different equation to calculate the frequency shift depending on whether the sound source or the observer is moving. However, if the speed of the object is small in relation to the speed of sound, the same simplified calculation principles can be applied to both cases. Thus, the frequency shift has one proportionality ratio both for the original frequency produced by the sound generator and for the speed of the observer or the sound source. On the other hand, the frequency shift is inversely proportional to the speed of sound in the medium. The simplified equation is therefore:

$$\text{Frequency shift } f_{\mathrm{d}}\,(\mathrm{Hz}) = \frac{\text{Original frequency } f_{\mathrm{o}}\,(\mathrm{Hz}) \times \text{Speed } V\,(\mathrm{m/s})}{\text{Speed of sound } C\,(\mathrm{m/s})}$$

The Doppler effect is certainly experienced in everyday life. For example, an observer will notice a change in pitch of the sound coming from a fast car as it passes the point of observation on a fast road. As the car approaches, the observer hears a sound which is higher pitched than the sound of the engine. Once the vehicle has reached the hearer and is moving away, the sound waves are 'stretched' and the observer hears the sound at a lower pitch.

The frequency also changes when the distance between the sound source and sound receiver does

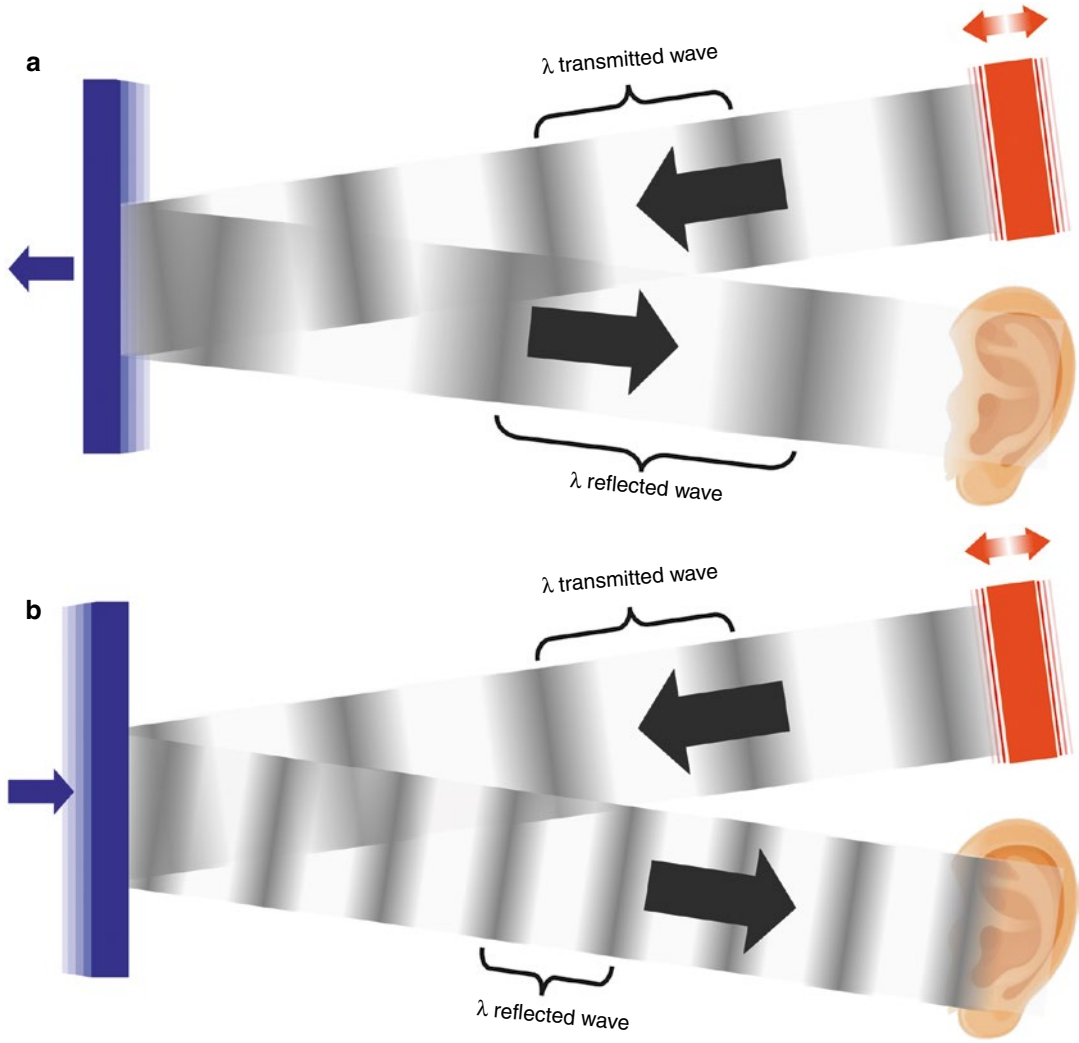

Fig. 1.4 (**a**, **b**) Change of the observed frequency when the reflector is moving away or towards

not change but the sound is deflected via a moving reflector (Fig. 1.4). Since the first Doppler shift has already taken place on arrival at the reflector and the second Doppler shift takes place when it is emitted at a frequency which has already changed, the frequency change is doubled.

In this case, the Doppler equation must be modified as follows:

$$\text{Frequency shift } f_d\,(\text{Hz}) = \frac{\text{Original frequency } f_0\ (\text{Hz}) \quad \text{Speed } V\,(\text{m/s}) \times 2}{\text{Speed of sound } C\,(\text{m/s})}$$

If the original frequency is known and the reflected frequency is measured, the speed of the reflector can also be measured by rearranging the equation. This principle is used in research, medicine and industry as well as the police force, military applications and by aerospace authorities to determine the approach speed of vehicles, aircraft or rockets. In these cases, the object being checked is not subjected to sound waves but high-intensity long-range electromagnetic waves.

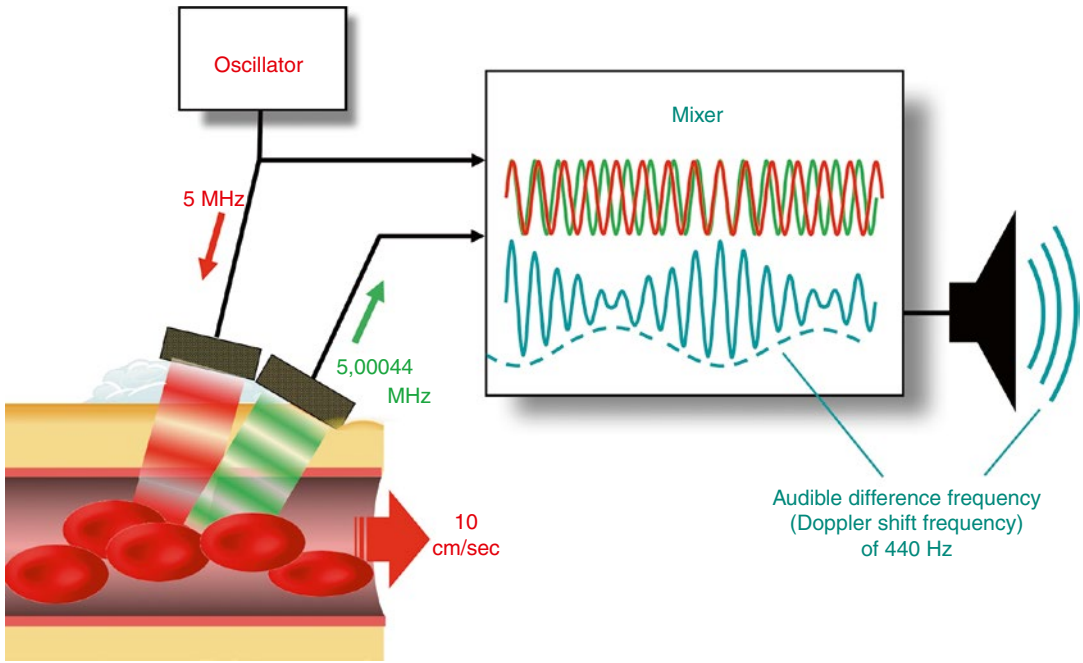

Fig. 1.5 Functional principle of a simple CW-Doppler without display

Different methods are available for Doppler sonographic flow measurements in medicine which will be dealt with in more detail in the following – the continuous wave Doppler, the pulsed wave Doppler and the different colour-coded Doppler techniques.

1.2 Continuous Wave Doppler Ultrasound Technique

The principle of frequency shifts caused by a moving reflector described above is used to measure blood flow velocity (Satomura 1959; Strandness et al. 1967).

Ultrasound is continuously radiated into the body at a frequency between 2 and 10 MHz from a quartz crystal (Fig. 1.5). Minute components of the energy are reflected back by the tissue, vessel walls and even the surfaces of the corpuscular blood components and picked up by a second crystal. The detected ultrasound is fed to a so-called mixer where it is compared with the transmission frequency. Since there is no difference between the frequencies of the sound reflected back from the static reflecting surfaces to the frequency of the transmitted sound, the mixer will not detect any difference between them.

The moving blood cells, however, reflect back the radiated sound at a different frequency according to the Doppler principle. The greater the rate of flow, the greater the difference between the frequency of the reflected sound and the frequency of the radiated sound. With a transmission frequency of 5 MHz and a velocity of 10 cm/s, the ultrasound frequency shifts from 5,00,000 to 5,00,044 MHz. To determine the difference, the transmitted and received frequencies are superimposed so that they alternate between coincidence and noncoincidence as a result of the slight difference in wavelengths. At the same time, the amplitude is always increased by addition when both signals are exactly in phase, i.e. have the same polarity. On the other hand, if the ultrasound frequency has a positive amplitude while the other signal is negative, the net result is extinction. This produces a new frequency which corresponds exactly to the difference between the transmitted ultrasound frequency and the received ultrasound frequency. In the example here, the difference is 440 Hz, which can be heard as a howl or a hiss in the loudspeaker.

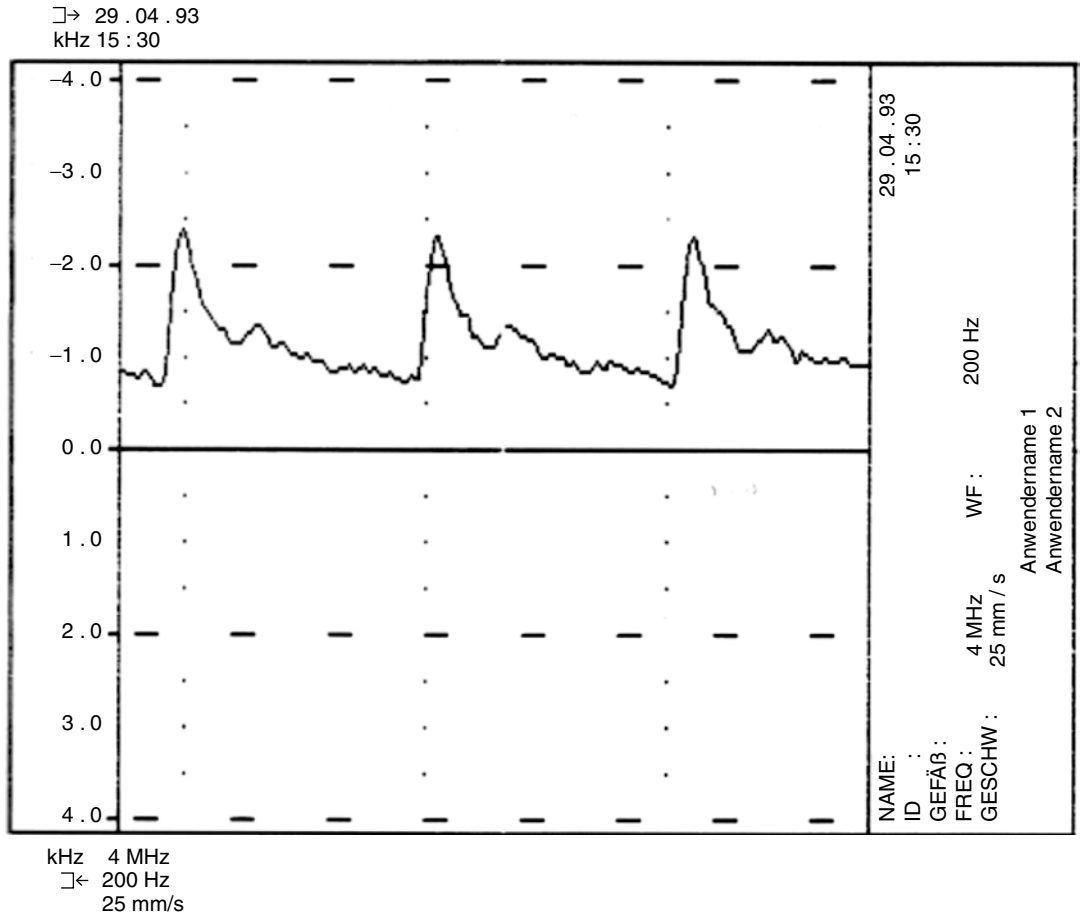

Fig. 1.6 Print out of a Doppler curve by application of a zero crosser signal processing

The faster the blood velocity, the greater the difference between the transmitted frequency and the received frequency and the greater the perceived difference or Doppler frequency

Since blood flows at an almost uniform rate in the veins, the noise coming from the device during a venal examination is an almost constant howl. On the other hand, the flow pulsation in the arteries causes the sound pitch to swell and subside as the Doppler frequency changes with the cardiac cycle.

Although this simplest of all Doppler devices only provides acoustic information about flow behaviours, it can be used for numerous applications in angiological diagnostics (Marshall 1984; Fischer and Wuppermann 1985; Mühlen 1989).

Since the acoustic signal from this device corresponds merely to the difference between the transmitted and received frequency, it is possible to draw conclusions on the flow velocity only – not on the flow direction. Both a blood flow towards the probe and a blood flow away from it but of identical flow velocity will produce an identical frequency shift. This technique is therefore called '*non-directional*'.

The Doppler systems which indicate the direction of flow have a wider range of applications. The signal processing system of these devices is able to detect the direction of flow in relation to the probe and display it as a velocity curve on the screen or printer (Fig. 1.6). Usually, these *directional Dopplers* are polarised so that flows towards the probe are displayed above and flows away from the probe are displayed below the zero line on the screen. However, this arrangement can be inverted

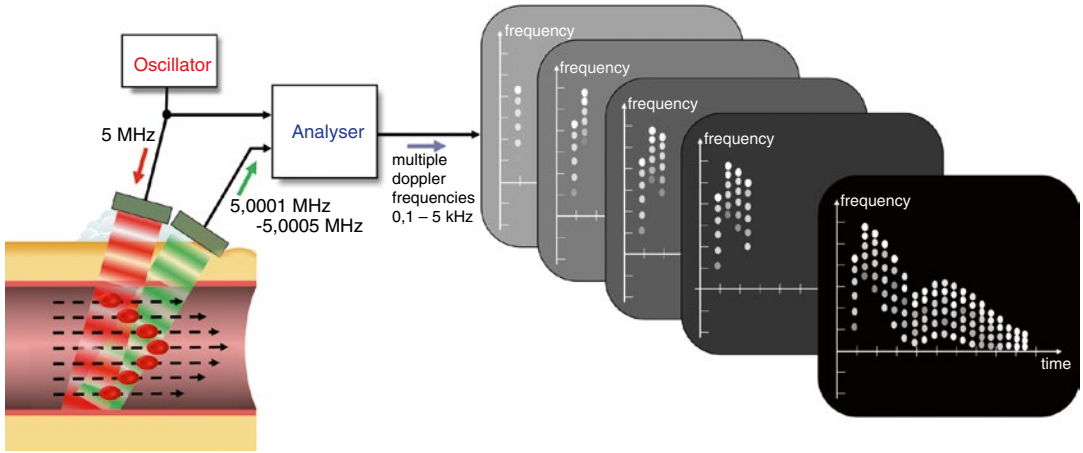

Fig. 1.7 Acquisition and display of a Doppler frequency spectrum

on every commercially available device. The size of the deflection, i.e. the amplitude of the curve, reflects the flow velocity under examination.

Simple devices use a so-called *zero crosser* to convert the Doppler frequency into a proportional deflection. Here, the frequency of the Doppler signal is simply 'counted out' and converted into an analogue deflection (MacLeod 1967).

These devices therefore do not take into account the fact that the corpuscular blood components move at very different rates over the cross section of the vessel. Since the zero crosser is not able to record several speeds simultaneously, only the predominant speed (i.e. the speed at which the overwhelming majority of particles are moving within the observation time) appears on the screen. All other speeds are disregarded (Lunt 1975; Evans et al. 1989).

However, since the evaluation during Doppler sonographic examinations predominantly relies on measuring the peak velocity (Gosling 1971; Gosling and King 1974; Planiol and Pourcelot 1973), applying zero crossing is out of the question for all quantitative and semi-quantitative analyses.

1.3 Spectral Analysis

In contrast to the zero-crossing technique, spectral analysis shows all the speeds over the cross section of the vessel (Maulik 2005). In addition to this, the frequencies or velocities on the screen are correct, so it is possible to make quantitative statements by including other measurement parameters (Warnking and Teague 1981). In principle, the velocity spectrum displayed on the screen or printer corresponds to the conventional Doppler curves with a vertical speed axis and horizontal time axis.

Figure 1.7 shows how it operates schematically. In order to show all the speeds simultaneously, the particle speeds detected are displayed on the vertical axis in the form of dots one on top of the other. Each dot therefore corresponds to a certain frequency or speed. Each speed causes its own Doppler shift so the receiving crystal detects a variety of different ultrasonic frequencies. However, since this receiving crystal cannot oscillate at all the frequencies simultaneously, they are superimposed additively to produce a new, quite complex curve shape. Inside the spectrum analyser, this frequency mix is compared with the original transmitter frequency to produce numerous different Doppler or difference frequencies. Thus, the slow velocity at the wall yields a Doppler frequency of a few Hz, while the highest speed which is often in the centre of the vessel can produce a frequency shift of up to 10 kHz or more, for example.

These Doppler frequencies are displayed on a vertical frequency axis on a screen in the form of pixels. Each dot therefore represents a certain frequency or speed. The intensity of each dot represents the occurrence of the corresponding speed. Thus, dots which are particularly intense indicate that there are a lot of blood cells moving at this speed.

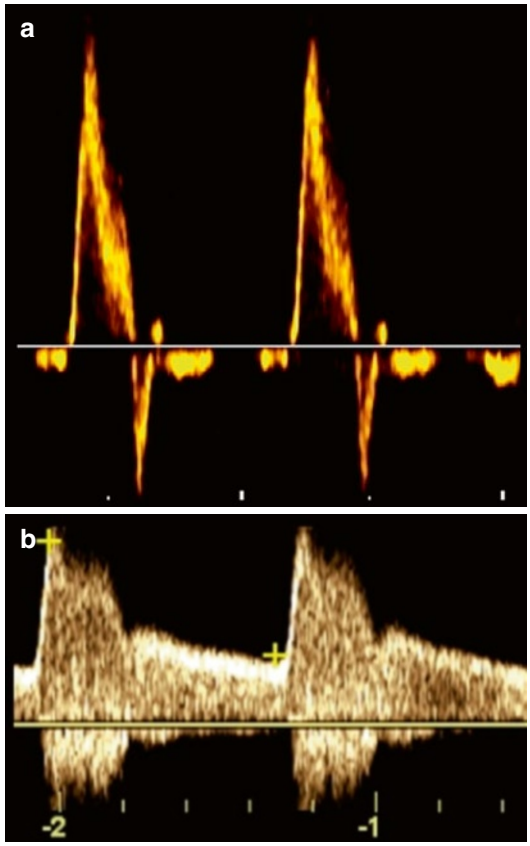

Fig. 1.8 (**a**) Frequency spectrum of an undisturbed laminar flow. (**b**) Frequency spectrum of a turbulent flow

The frequency analysis and the structure of the corresponding vertical dot line are completed after a few milliseconds. A new measurement cycle then starts immediately where the frequencies which are now determined are displayed to the right of the previous line. Thus, the picture develops line-by-line in quick succession as the speed curve is displayed to the observer and scrolls across the screen in real time.

Under normal conditions, the blood flow is predominantly laminar, i.e. the corpuscular blood components flow in a straight or – as we now know – in a helical (Kilner et al. 1993) way on parallel paths. In this case, the flow rate profile is plug shaped, i.e. most particles move at approximately the same speed – only the particles in the direct vicinity of the wall move at a lower speed due to friction. As shown in Fig. 1.8a, the spectrum analyser in this case displays a narrow-band frequency curve. Low speeds are underrepresented, particularly during the systolic phase, and the so-called systolic window appears, i.e. an almost dot-free area below the ends of the curve envelope. The dots representing the higher speeds are particularly intense on the display.

Sometimes, particularly when higher speeds are displayed on the screen due to stenosis, the laminar nature of the blood flow can be substantially disrupted. In this case, the particles no longer move along straight parallel paths but change their paths and speeds; sometimes, eddies and other disturbances occur. Since this type of turbulent flow produces many different speeds simultaneously, the Doppler spectrum appears as a broad band during the systolic phase, and the curve between the zero line and the peak value is relatively uniformly filled with dots (Fell et al. 1981) (Fig. 1.8b). Components of the flow moving in opposite directions as they occur inside the eddies appear below the zero line on the Doppler spectrum.

Since the amplitude of the curve on the spectrum is exactly proportional to the actual peak velocity, this method makes it possible to carry out at least semi-quantitative evaluations, in order to calculate and use indexes, for example. The Resistance Index according to Pourcelot (Planiol and Pourcelot 1973) has proved particularly effective for most applications. This index makes it possible to draw conclusions about the perfusion resistance downstream of the measurement site by comparing the height of the systolic peak flow velocity with the end-diastolic peak flow velocity. The equation is as follows:

$$\mathrm{RI} = \frac{S - D}{S}$$

where 'S' is the systolic peak frequency and 'D' is the highest end-diastolic frequency.

Figure 1.9a shows typical flow-velocity curves as a function of the subsequent perfusion resistance and the RI calculation. A low peripheral resistance is accompanied by a high diastolic amplitude and low RI. On the other hand, a high peripheral resistance is characterised by low diastolic amplitude and a high RI (Fig. 1.9b).

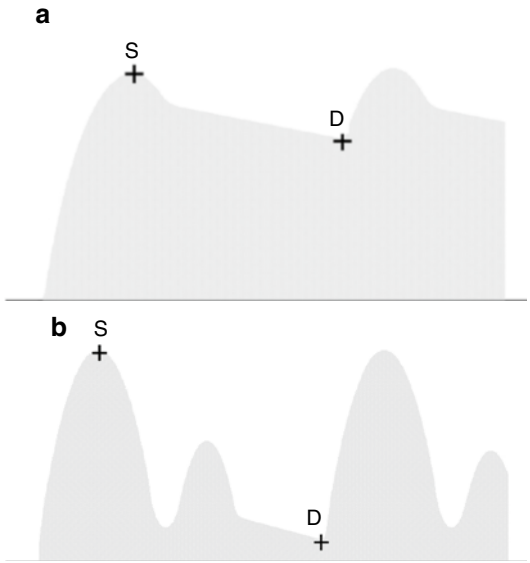

Fig. 1.9 (**a**) Schematic flow curve of a vessel supplying an area with low perfusion resistance. (**b**) Schematic flow curve of a vessel supplying an area with high perfusion resistance

Modern Doppler systems can detect the relevant measurement points S and D automatically and display the index on the screen in real time.

One alternative to the Resistance Index is the somewhat more complex Pulsatility Index of Gosling (Gosling and King 1974; Gosling et al. 1971). In contrast to the RI, this index not only detects the systolic and the end-diastolic peak but also takes into account the entire curve for the cardiac cycle, including any retrograde flow components which may be present. The peak-to-peak frequency must also be determined, i.e. the difference between the maximum and minimum frequency, where the minimum may also lie below the zero line in the case of an early diastolic retrograde dip (Fig. 1.10). Furthermore, an average value for the envelope must be calculated over one cardiac cycle – and this requires an electronic calculation.

The Pulsatility Index is produced from the ratio of the peak-to-peak value to the average value.

$$\mathrm{PI} = \frac{\text{Peak-to-peak frequency}}{\text{Time-averaged maximum frequency}}$$

Note: Some ultrasound systems offer different algorithms for determining the average value, e.g. by recording the integral, i.e. of all the frequencies beneath the envelope curve (TAM – time-averaged mean frequency). The PI values determined using these algorithms do not correspond to the measurement results based on the envelope curves. This must be taken into consideration during the evaluation and when comparing with the standard values.

1.4 Angle Problem

As already explained above and shown by the Doppler equation, a proportional relationship exists between the blood velocity and the Doppler frequency or the curve amplitude on the screen. However, this proportionality does not allow the determined Doppler frequency to be converted directly into the absolute blood velocity in m/s or cm/s.

This is due to the strong dependence of the Doppler frequency on the angle between the ultrasound beam and the direction in which the reflectors move. It is not the absolute speed of a reflector which is crucial for the development and intensity of the Doppler effect, but the velocity at which the distance between the reflector and the Doppler scan head changes. The difference between the absolute and detected velocity is illustrated in Fig. 1.11. While the erythrocyte moves at 1 cm per second in relation to its environment, its change in distance to the Doppler probe is smaller. For example, if the initial distance is 10 cm, the final distance after 1 s is only reduced to 9.3 cm. Therefore, from the point of view of the Doppler scan head, the approach velocity of the erythrocyte is merely 0.7 cm/s which means that the measured velocity is 30 % less than the actual velocity.

The angle between the direction of movement and the direction of observation is crucial for the difference between the measured velocity and the actual velocity – the wider the angle, the greater

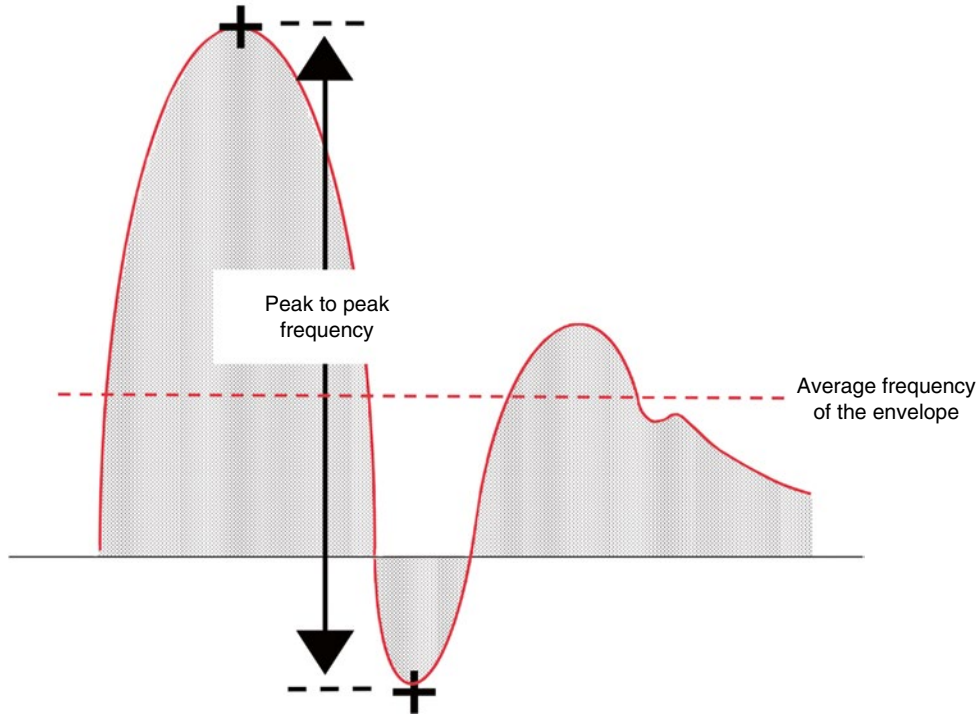

Fig. 1.10 Measuring points for calculation of the pulsatility index

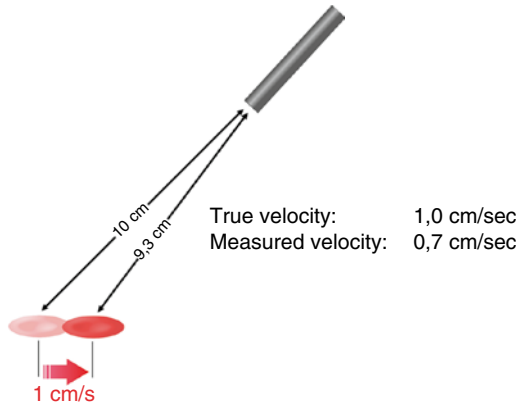

Fig. 1.11 Difference between true velocity and approach velocity

the discrepancy between the actual forward movement and the measured approach velocity (Fig. 1.12).

With angles up to approximately 20°, the discrepancy is small and the detected Doppler frequency only deviates a little from the value which would be attained under the ideal conditions of 0°.

The deviation is calculated very easily from the cosine of the angle. At 0° (when the erythrocytes move directly towards or away from the probe), the cosine = 1 so that the genuine frequency shift occurs according to Doppler's equation. At an angle of 15°, the cosine decreases to 0.97. Since this value enters the Doppler equation as a multiplication factor, the Doppler frequency which occurs drops to 97 % of the original value. The corresponding measurement error of 3 % can of course be disregarded. Even at 30° and a corresponding cosine of 0.87, the deviation amounts to a tolerable 13 % and is therefore still within the usual biological scatter of Doppler sonographic standard values. At 45°, the error increases to 25 % and at 60°, to a significant 50 % rising quickly as the angle widens. At 75°, the cosine is only 0.26 and the Doppler frequency detected by the equipment is therefore only 26 % of the frequency which would be produced under the optimum angle conditions. As the angle increases further to 90°, the cosine finally decreases to the value '0', so the Doppler frequency detected is only 0 % of the optimum

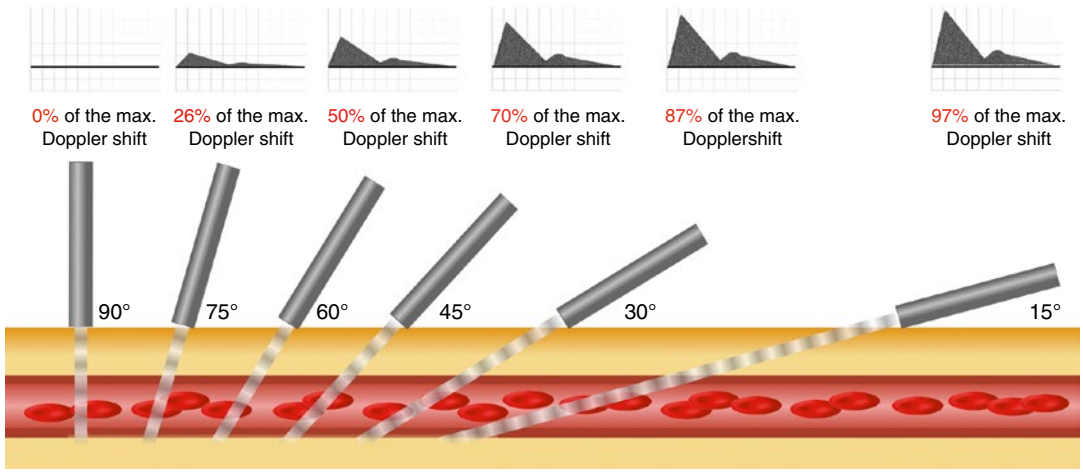

Fig. 1.12 Angle dependency of the Doppler frequency

value. The erythrocytes pass under the ultrasonic beam at right angles so they neither approach nor move away from the probe and no longer give rise to a Doppler effect. The spectral display on the screen usually only shows weak interference signals in the region of the zero line, and the Doppler signals which are heard are mainly caused by movements of the vessel wall.

As the matters discussed previously show, the examining physician should always endeavour to obtain an acute-angled beam of the target vessel and place the probe accordingly in order to produce the optimum signal. Experts largely agree that angles wider than 60° cannot be relied on to produce a sufficiently large Doppler shift and the frequency spectra they produce should not be assessed. If the angle which can be obtained is between 30° and 60° due to the anatomical conditions, the question how to continue the procedure depends on the objective of the examination and must be determined by the following criteria:

- With a purely qualitative evaluation of the frequency spectrum, particularly when the Resistance or the Pulsatility Index is used for the evaluation, the differences in frequency caused by the angle are immaterial since the systolic peak frequency is falsified to the same degree as the diastolic peak. The frequency spectrum displayed can therefore be used for the evaluation without further measures.
- On the other hand, if the aim is to acquire a quantitative measurement of the flow velocity, the angle between the ultrasonic beam and the axis of flow must be known so that the cosine of this angle can be used for conversion into m/s or cm/s.

$$\text{Flow velocity}\,(\text{m/s}) = \frac{\text{Doppler frequency}\,(\text{Hz}) \times \text{Speed of sound}\,(\text{m/s})}{\text{Ultrasound frequency}\,(\text{Hz}) \times 2 \times \cos\,\theta}$$

Because Doppler and B-mode imaging sonography are normally used in combination today, the angle can be seen directly on the screen by manually adjusting the angle cursor along the direction of flow visualised by the B-mode sonography. The system computer therefore receives the necessary information about the angle between the ultrasound beam and the axis of flow, determines the cosine automatically and takes this value into account when outputting the data and scaling the Doppler curve (see also the Sect. 1.6, Fig. 1.20).

If it is possible to use an acute angle of 30° or even narrower, there is usually no need for correction in the case of quantitative measurements

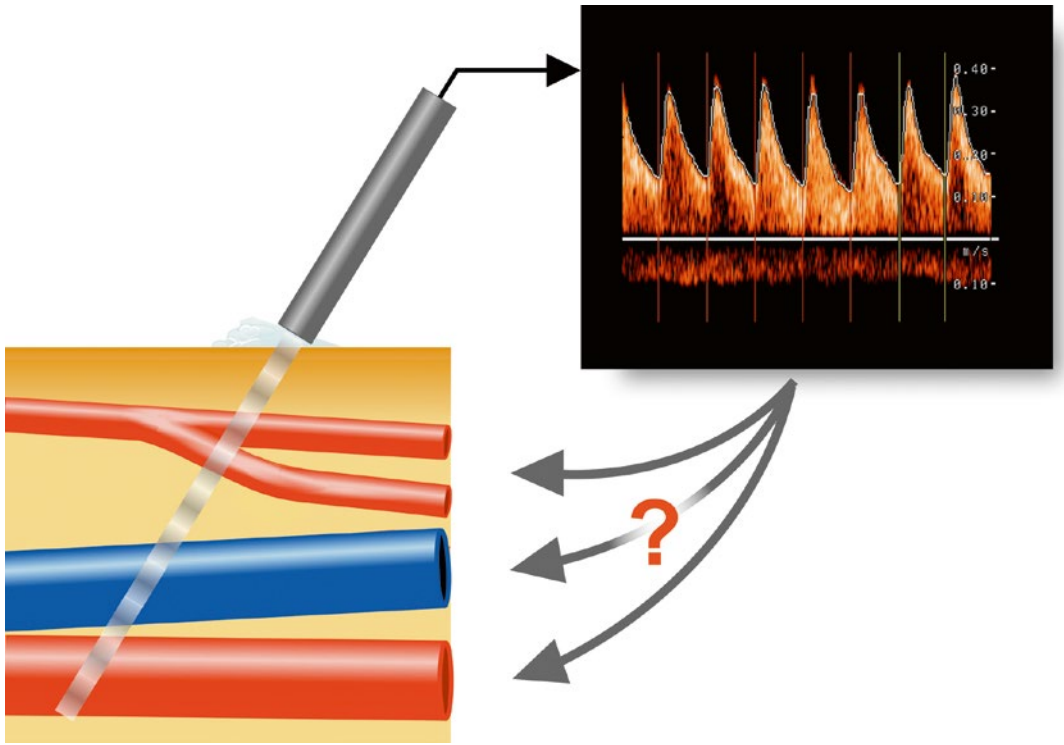

Fig. 1.13 Missing possibilities of depth selection by using CW Doppler

since the Doppler frequency detection error can be neglected (underestimation of 13 %).

1.5 Pulsed Doppler Devices

When there are other blood vessels between the Doppler probe and the vessel under examination, the *continuous wave technique* (CW Doppler) described above will not work because it is unable to target the depth. Continuous wave devices with their continuous sound transmission detect all the blood flows present within the range of the ultrasonic beam (Fig. 1.13).

With pulsed Doppler devices, however, blood vessels can be selected for measurement within a range of depths which can be chosen beforehand (Baker 1969).

Pulsed Doppler systems do not transmit continuous ultrasound but wave packets of short duration (bursts) lasting just a few microseconds each. Immediately after a transmission pulse has been transmitted (Fig. 1.14a), the crystal in the ultrasonic probe is switched to the receiver in order to detect the sound reflected back (Fig. 1.14b). However, not all the reflected signals are processed, but only those which are reflected back at a certain time and, therefore, from a certain depth (Fig. 1.14c). Echoes which arrive too early, i.e. from close range, and signals which return very late, i.e. from greater depths, are ignored (Fig. 1.14d).

The time window and, therefore, the depth where the measurement is to take place can usually be varied using a suitable controller, 'blind' either by means of a device with a cm scale or by placing a cursor inside a reference ultrasonic image on the screen.

In most cases, the length of time the receiver is switched on can be changed so that the axial length of the measurement site can be varied. However, the lateral dimension depends on the diameter of the ultrasonic beam alone and cannot be varied with the latest technology. Since the volume examined represents a three-dimensional space, it is frequently also referred to as 'sample volume'.

Once the specified waiting period has elapsed and the expected echoes have been received, the crystal can be used again to send out the next transmission pulse. The frequency at which these transmission pulses are sent, the so-called *pulse*

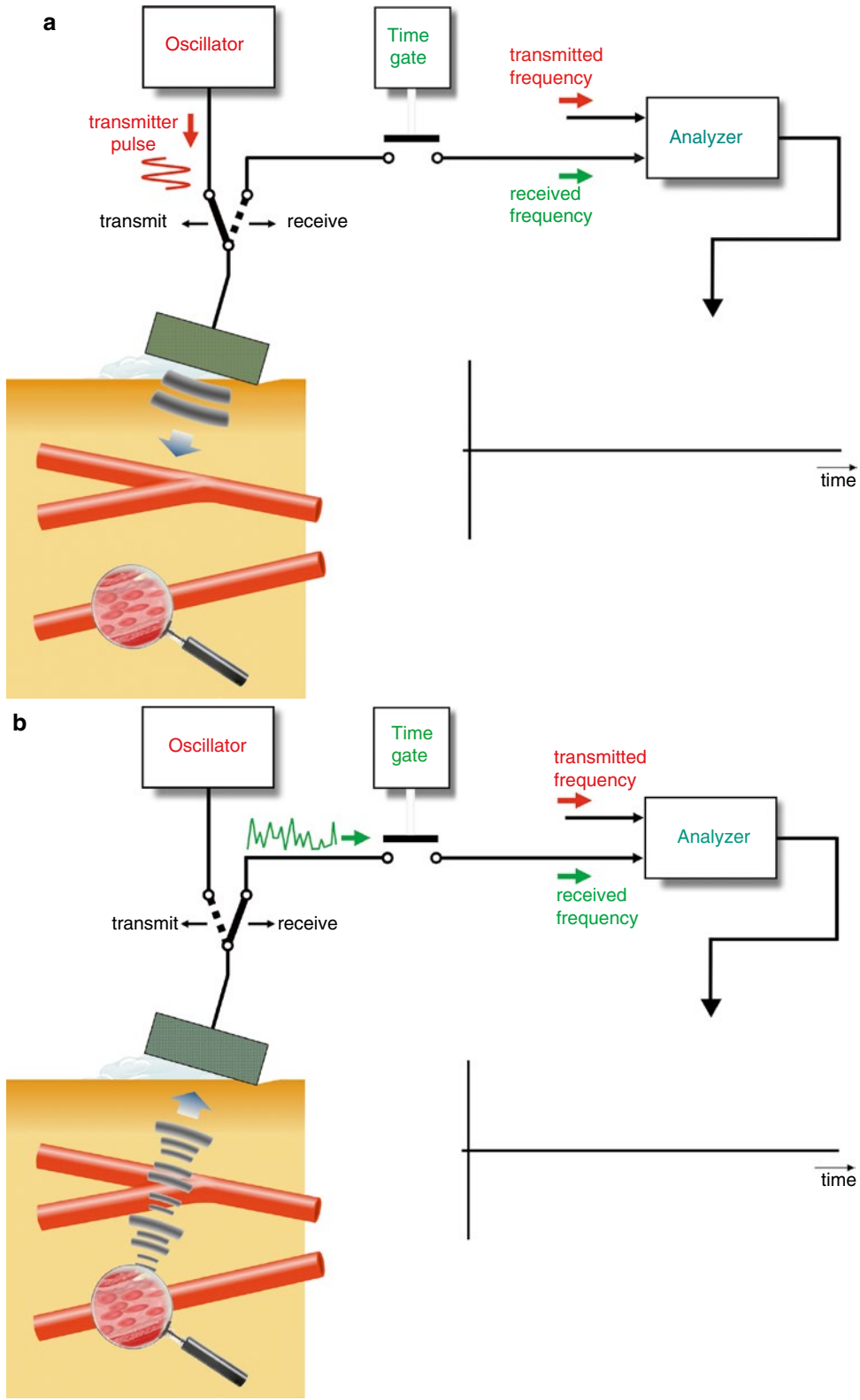

Fig. 1.14 Functional principle of the PW Doppler. (**a**) Transmission of the ultrasound burst. (**b**) Switch to reception whereas unwanted signals from the near field are ignored. (**c**) Time gate switches to admission, signals coming from the target area are processed. (**d**) Time gate switches to block, signals from greater depths are ignored

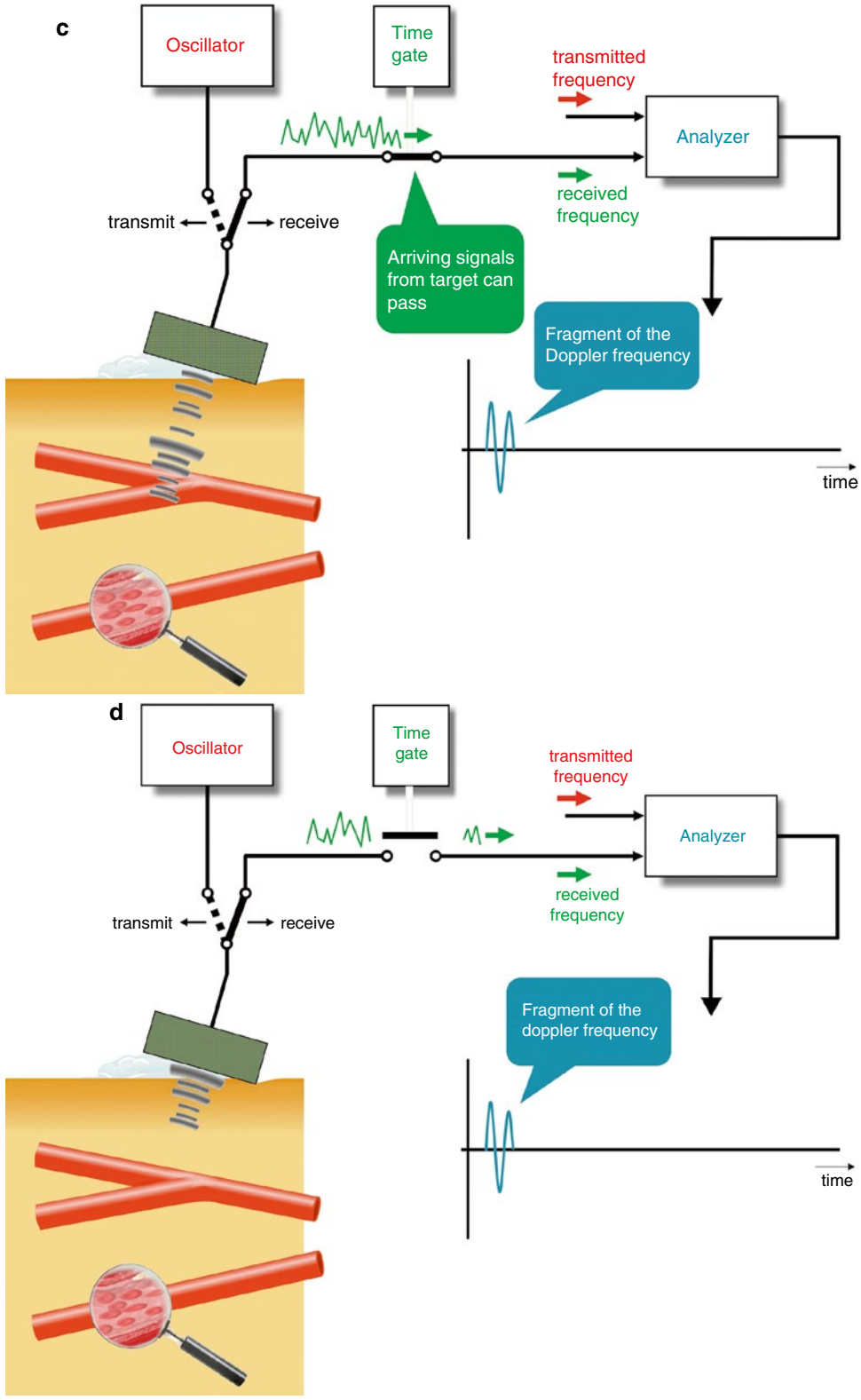

Fig. 1.14 (continued)

repetition frequency (PRF), therefore depends on the waiting period and thus the required measurement depth. For example, if the measurement should take place at a depth of 5 cm, the period between transmission and reception is 0.06 ms and the resulting PRF must not be greater than approximately 16.6 kHz. On the other hand, for a measurement depth of 20 cm, for example, wave propagation back and forth takes 0.24 ms so the PRF in this case can only be approximately 4.15 kHz at the most.

Since the ultrasound is not continuous but is generated in the form of very short pulses, the reflected ultrasound frequency also only occurs at intervals. The Doppler frequency is therefore also not available in the form of a continuous wave after the mixer but is made up of separate segments with gaps in between. If the PRF is high in relation to the Doppler frequency, then the wave shape of the Doppler frequency can be reconstructed unambiguously from the separate pulses (Fig. 1.15 top). On the other hand, if the PRF is low because the

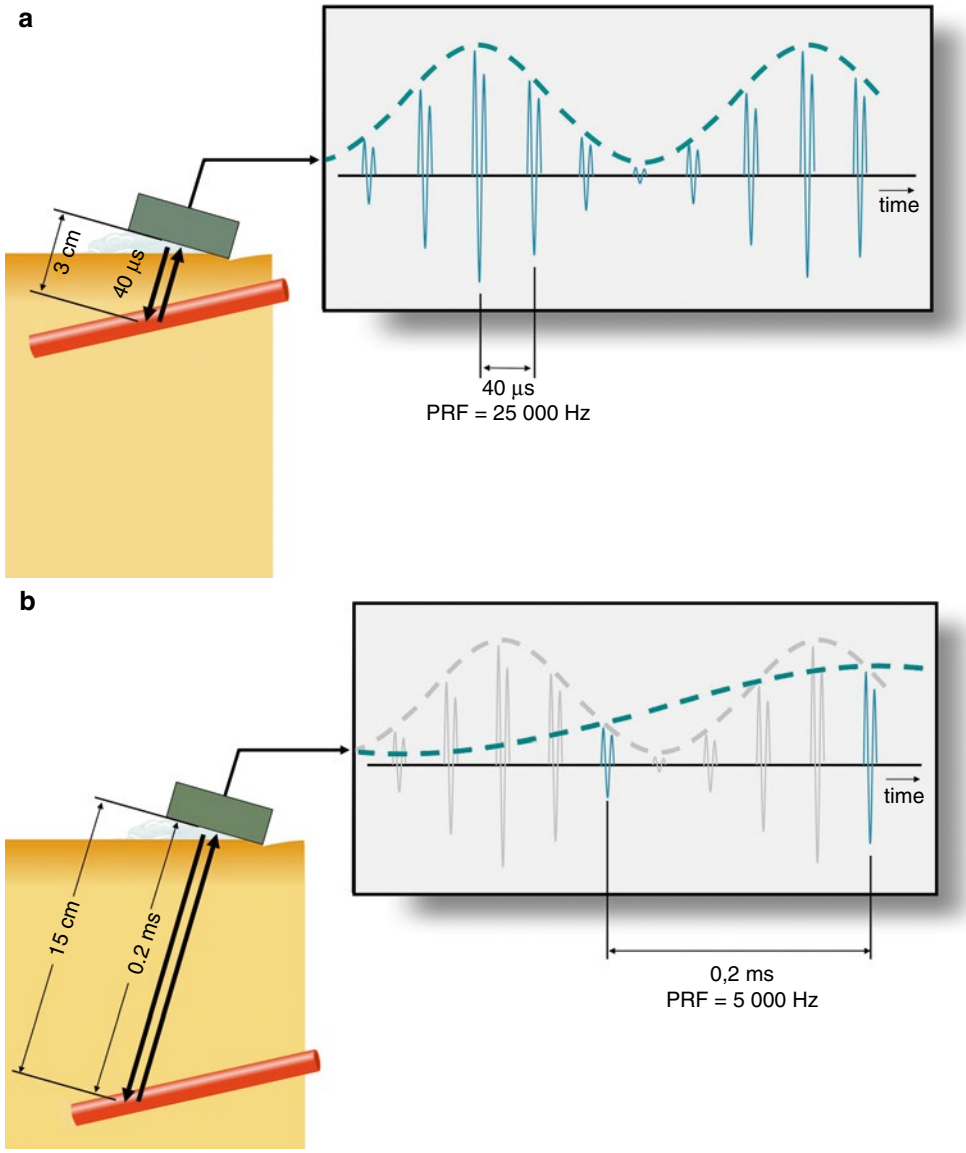

Fig. 1.15 Dependence of the PRF from sample volume depth. Measurement error caused by low PRF

sample volume is at a distance and the time interval between two consecutive pulses is therefore long, there may not be enough separate pulses available to be able to recognise the wave shape of the Doppler curve (Fig. 1.15 bottom).

As the lower part of Fig. 1.15 shows, the frequency analyser has reconstructed the curve incorrectly in this case to produce a Doppler frequency which is too low (Smith 2001).

To avoid producing measurement errors, the PRF must be at least twice as large as the maximum Doppler frequency being measured.

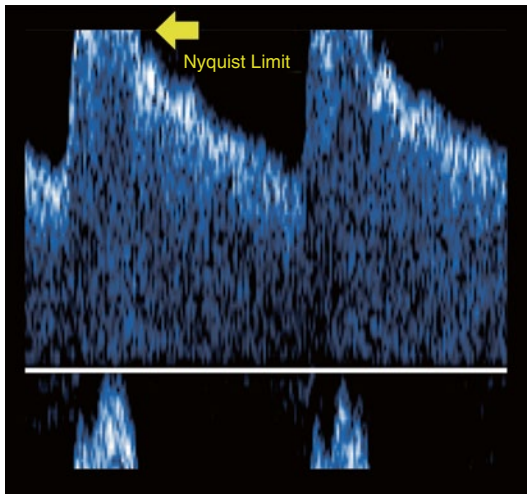

Fig. 1.16 Aliasing due to low PRF

The maximum measurable (PRF dependent) Doppler frequency is called the *Nyquist limit*. Once the velocity in the vessel under examination exceeds the Nyquist limit, there will usually be significant measuring errors, as described. The Doppler frequencies which occur beyond the Nyquist limit usually appear below the zero line (Fig. 1.16).

Cutting off the frequencies above the Nyquist limit (displaying these signals below the zero line) is known as *aliasing*. The measurement range can be increased and the aliasing phenomenon overridden by moving the zero line (Fig. 1.17). Here, removing the aliasing is purely virtual in nature since the (aliasing) pixels which appear on the lower edge of the scale are displaced and displayed on the upper edge of the scale.

In extreme cases, when the zero line is placed on the upper or lower edge of the scale, sometimes only unidirectional flows can be displayed. However, the zero shift function only allows the measurement range to be doubled; if doubling is not sufficient, the PRF must be increased. Since the PRF has a direct influence on the velocity or frequency measurement range, the scale of the Doppler curve displayed also changes as the PRF is adjusted (Fig. 1.18). It therefore does not make any sense to select a particularly high PRF when deriving a low-frequency Doppler spectrum. Because the measurement range is wide in this case, the Doppler curve displayed would be much too small and the interpretation more difficult.

However, the adjustment range of the PRF is also limited (as already explained), particularly

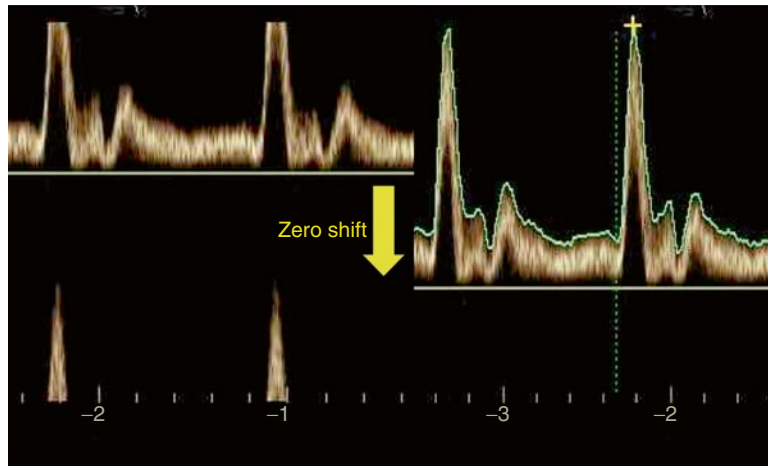

Fig. 1.17 Elimination of the aliasing artefact by shifting the zero line

Fig. 1.18 Relation between PRF and velocity range

while scanning deep vessels; the time interval between sending a transmission signal and receiving the corresponding echo complex involves problematic values. Since a transmission signal can only be sent out when the echoes of the preceding pulse have returned, the maximum possible PRF is inversely proportional to the depth of the sample volume.

Some Doppler devices have a *high PRF* mode. With this technique, the examination is carried out with a pulse repetition frequency which is too high for the examination depth by an exact factor of 2, 3 or 4. Thus, 1 or 2 or 3 additional secondary measurement volumes form between the ultrasonic probe and the desired measurement depth. The Nyquist limit is shifted upwards by the corresponding amount simultaneously.

During the examination, it is only necessary to make sure that there are no vessels located in the additional sample volumes which could falsify the measurement result.

1.6 Duplex Systems

Pulsed Doppler devices are usually used in combination with ultrasonic B-mode image devices, the so-called *duplex systems*, in order to adjust the size and depth of the sample volume correctly. In order to meet the needs of a wide variety of different applications, the industry now offers various different solutions – in all cases, however, the sample volume can be positioned under visual control to ensure that the vessel under examination can be selected with sufficient precision.

The mechanical sector systems used previously only provided mode switching, meaning that the simultaneous presentation of a real-time ultrasonic B-mode image and Doppler spectrum was of course not possible.

Today's duplex systems predominantly operate with array ultrasonic probes (Fig. 1.19). Unlike the mechanical method, these make it possible to produce the scanning lines of the B-mode image and the Doppler line in very quick succession. Thus, the first image line is produced initially and then the device switches to Doppler mode and the second image line follows. After this, the device switches back to Doppler mode again and so on. By switching between the two modes very quickly, the examining physician sees a genuine simultaneous image of the moving two-dimensional B-mode image and the Doppler spectrum. In this case, every crystal in the array can be involved to produce the Doppler beam so that it can be swung freely over the field of view. The beam direction of

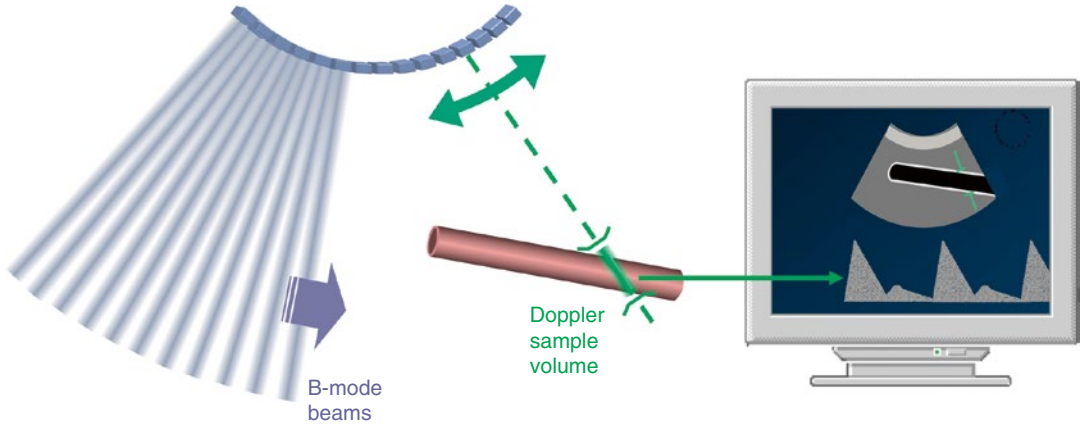

Fig. 1.19 Functional principle of the duplex mode

the Doppler and the position of the sample volume are simulated and superimposed on the ultrasound B-mode image. A small reference line can also be displayed which the examining physician aligns manually with the vessel or the direction of flow using a track ball or rotary control (Fig. 1.20). Thus, the system is informed of the angle between the Doppler beam and the direction of flow, and the scale and the Doppler curve are given absolute dimensions in cm/s or m/s.

Note: The scale for the Doppler curve is only calibrated and can only be consulted for quantitative statements about the flow rate once this angle cursor has been activated and correctly adjusted.

Note: The angle cursor is used exclusively for calibrating the velocity scale. However, it does not allow unsuitable, wide scanning angles above 60° to be used under any circumstances.

If it is only a semi-quantitative measurement which is to be carried out (e.g. for determining the Resistance Index), the angle cursor does not need to be used since it is only the relationship between the systolic and diastolic amplitude which is measured.

1.7 Colour-Coded Doppler Sonography

It has already been explained in detail how pulsed Doppler operates at selected depths by using a time window so that only those echo signals which come back at a certain point in time and therefore from a certain depth can be processed.

By modifying the time gate, it would also be quite possible to construct several sample volumes along the axis of the beam. However, in this case, several display or registration channels would have to be available so that the pieces of speed information reflected back in succession from the different sample volumes are not superimposed one on another but are kept separate. This type of multichannel Doppler has actually been built in the form of laboratory setups and has provided research with new information about the flow behaviour of blood. By using this method, it was possible to distribute several sample volumes over the diameter of a vessel and to determine the different velocities over the vessel cross section for the first time (McLeod 1974; Keller et al. 1976; Hoeks 1982).

This type of multichannel Doppler could be further modified by setting up several scanning lines next to one another, each consisting of multiple sample volumes. This would involve moving a single piezoelectric crystal mechanically or having to consult an array ultrasonic probe. The equipment would supply not only information about the flow conditions along an axis but also

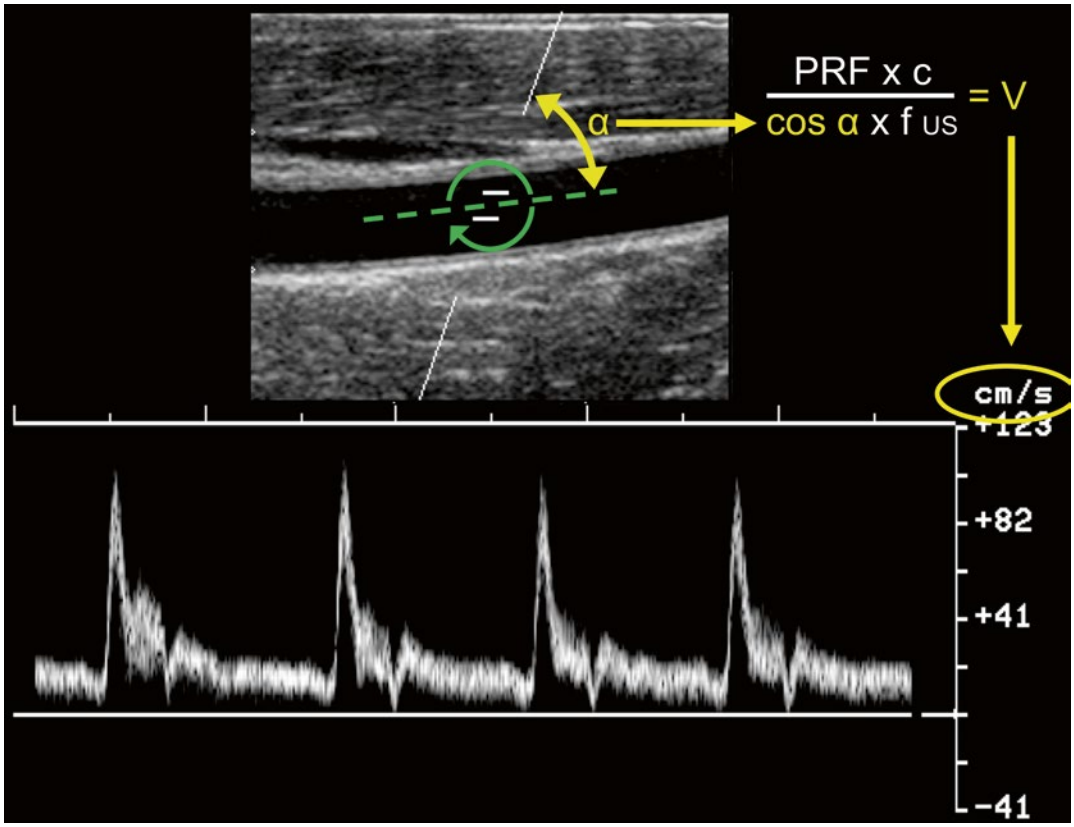

Fig. 1.20 Insertion of the angle cursor to calculate the cos alpha

selective information about all the flows occurring in a large area. However, since all the flow information would be present in isolated form, many hundred display or registration channels would be needed. On the one hand, such huge technical elaboration is out of question, but on the other hand, two-dimensional Doppler would expand diagnostic possibilities. American and Japanese developers have independently found a completely new and less complex way to display 2D flow data: the colour-coded Doppler method (Bommer and Miller 1982; Namekawa et al. 1982; Omoto 1984).

It is characteristic of this method that a pixel on the monitor is assigned to each sample volume, the pixel corresponding exactly to the sample volume in relation to its position and size. When movements of a reflector occur inside a sample volume, then the associated pixel is displayed in colour. If the reflected frequency produced in the sample volume is higher than the transmitter frequency (flowing towards the probe), the corresponding pixel turns red. On the other hand, blood which is flowing away from the ultrasound transducer appears as a blue pixel on the display (Fig. 1.21).

Not only the direction of flow but also the velocity of flow can be displayed and are expressed qualitatively by the brightness of the pixels. A high approach speed therefore produces a light red spot and a low approach speed (or an wide angle between the Doppler beam and direction of flow) produces a dark red spot. The same applies to the flow in the reverse direction for the colour blue. Here, it is essential to understand that it is not the peak but the average velocity which is expressed by a bright red.

If the corpuscles move inside one sample volume at very different speeds and in different directions, then a green point is often displayed in the corresponding position on the screen to indicate a disturbed flow with eddy currents.

Today's systems, however, are not limited to displaying reflections from the blood corpuscles

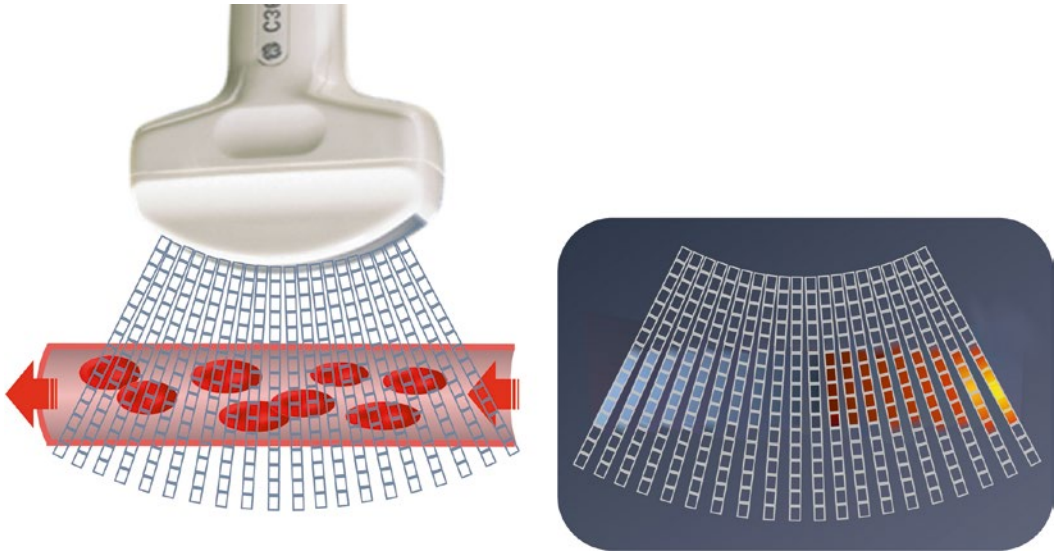

Fig. 1.21 Functional principle of the color-coded Doppler mode

which cause a Doppler effect. The comparatively strong echo signals from the tissue are also detected in the usual way and displayed on the screen in corresponding grey shades. However, since other shorter transmitter signals than those for Doppler sonography are beneficial for a high-quality B-image, the two modes are not generated in exactly the same instant. Actually, the B-mode module and the colour Doppler operate in very quick succession so the examining physician is given the impression of a simultaneous, real-time image. In comparison to the conventional B-mode image, the lumina of blood vessels are no longer black but appear red or blue depending upon the relative direction of flow (Fig. 1.22). However, it is essential that the examining physician takes into account that, in places where the angle between the blood flow and ultrasonic beam is 90°, no Doppler effect can take place ($\cos \theta = 0$). In this case, the screen stays black despite the fact that there is flow. Also, the colour-flow imaging system does not recognise the actual, absolute direction of flow but only the direction of flow in relation to the sample volume concerned.

This obvious restriction in expression has particular implications when the sector or convex method is used for the scanning. Since the acoustic scanning lines fan out from the scan head with this type of ultrasonic probe, the flow crosses the sample volumes at different angles when examining a straight vessel. Even if the flow velocity is invariable throughout the entire length of the vessel under examination, the corresponding colour display varies greatly. In the Fig. 1.22 example, the flow through the sample volumes on the outside right is at a very sharp angle (high approach speed) and the resulting pixels appear light red. Further to the centre, the angle between the scanning beam and the flow gets wider, which corresponds to a lower approach speed. The colour pixels therefore appear dark red.

Directly beneath the ultrasonic probe, the angle of flow through some sample volumes is at an angle of 90°. In this case, the reflecting corpuscles merely flow past the ultrasonic probe and there is no flow towards the probe. The corresponding image lines therefore also remain dark.

On the left side of the field of view, all the corpuscles are moving away from the ultrasonic probe, and so this part of the vessel lumen appears blue on the screen. As with the behaviour on the right side, the flow through the sample volumes on the very outside is at a very sharp angle (corresponds to the colour light blue), while the angles more to the centre are wider (corresponds to the colour dark blue).

When using the linear array method shown in Fig. 1.23, the problem described above does not

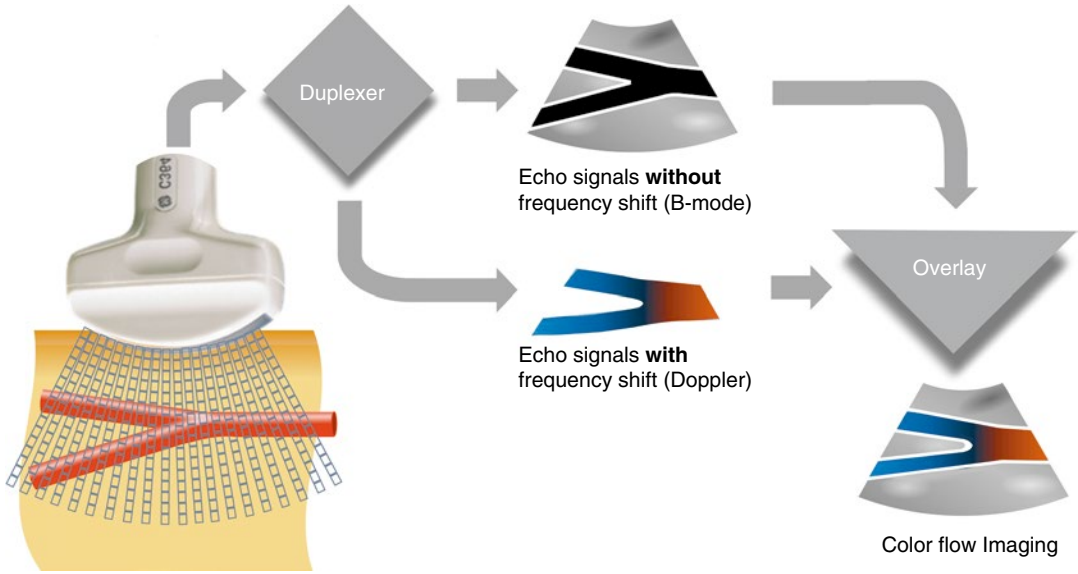

Fig. 1.22 Superimposing of color-coded Doppler signals with B-mode echoes

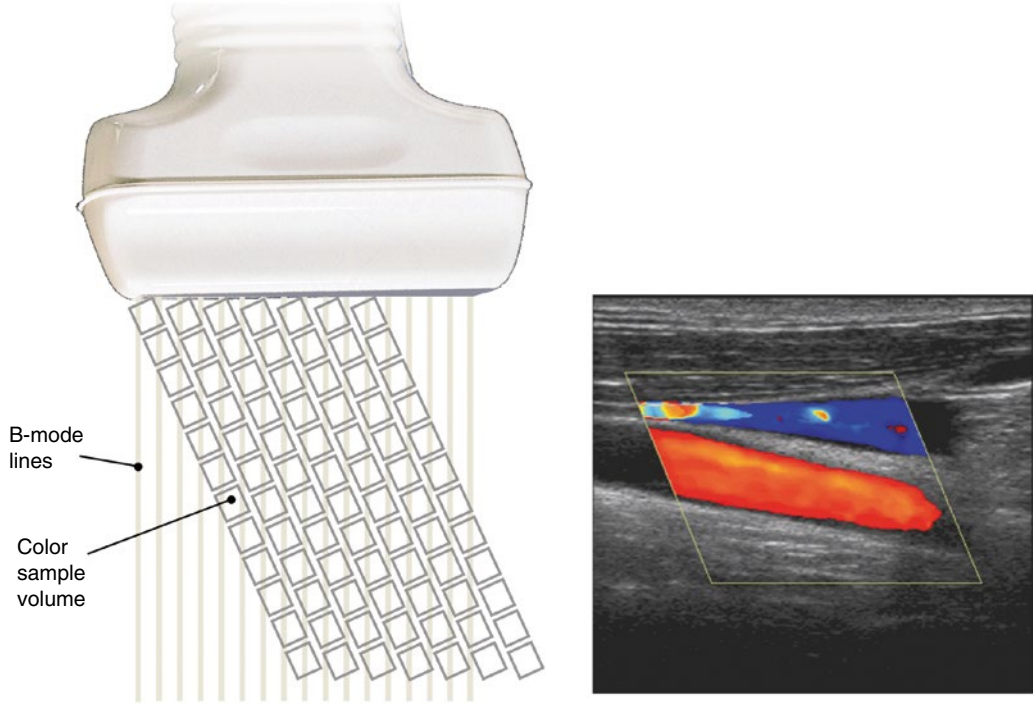

Fig. 1.23 Color-coded Doppler in conjunction with the linear array scanning method

occur. While the scanning lines for the 2D image develop at right angles to the surface of the ultrasonic probe, the radiation for the Doppler scanning lines is at a sharper angle. All these Doppler lines are sent out at the same angle and, therefore, in parallel. Thus, when using this method, no system-dependent colour irregularities should be expected.

The Doppler scanning lines are angulated by triggering the different transducer elements of the

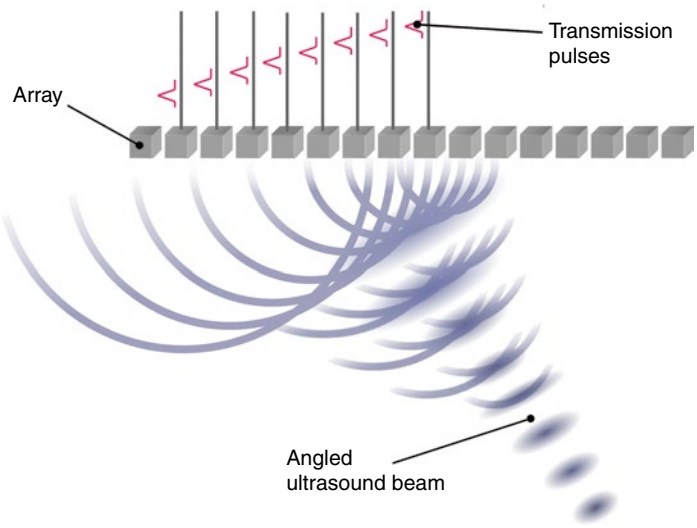

Fig. 1.24 Angulation of ultrasound beams through time-shifted triggering of the appropriate crystals

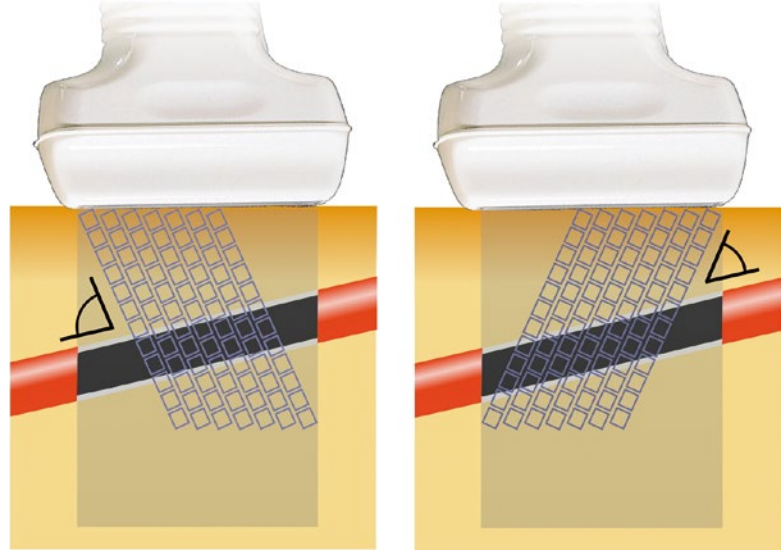

Fig. 1.25 Swivel of the color window to reach an optimal angle

group which are active and assigned to a scanning line at slightly different points in time in each case. Overlaying the individual sound fields produces a beam with an axis which can deviate from the vertical by up to approximately 30° depending on the time offset of the transmission signals (Fig. 1.24).

The main area of use for this linear array method is in the examination of vessels which run near to and parallel to the surface of the skin. Here, the angulation of the scanning lines and therefore the so-called colour window must be adjusted so that the angle between the axis of flow and the scanning sweeps is as sharp as possible (Fig. 1.25).

As already explained, when the measurement range of a pulsed Doppler system is limited as soon as the Doppler frequency exceeds half of the pulse repetition frequency (the frequency at which transmitted pulses are beamed into the body), the corresponding frequency points on the Doppler spectrum appear on the neighbouring channel. This effect, known as aliasing, also affects the colour-coded system at high flow rates. In this case, the aliasing occurs in the form of a colour change. Thus, a flow towards the ultrasonic probe with a Doppler frequency which exceeds half of the PRF will appear on the screen incorrectly as blue instead of red, for example.

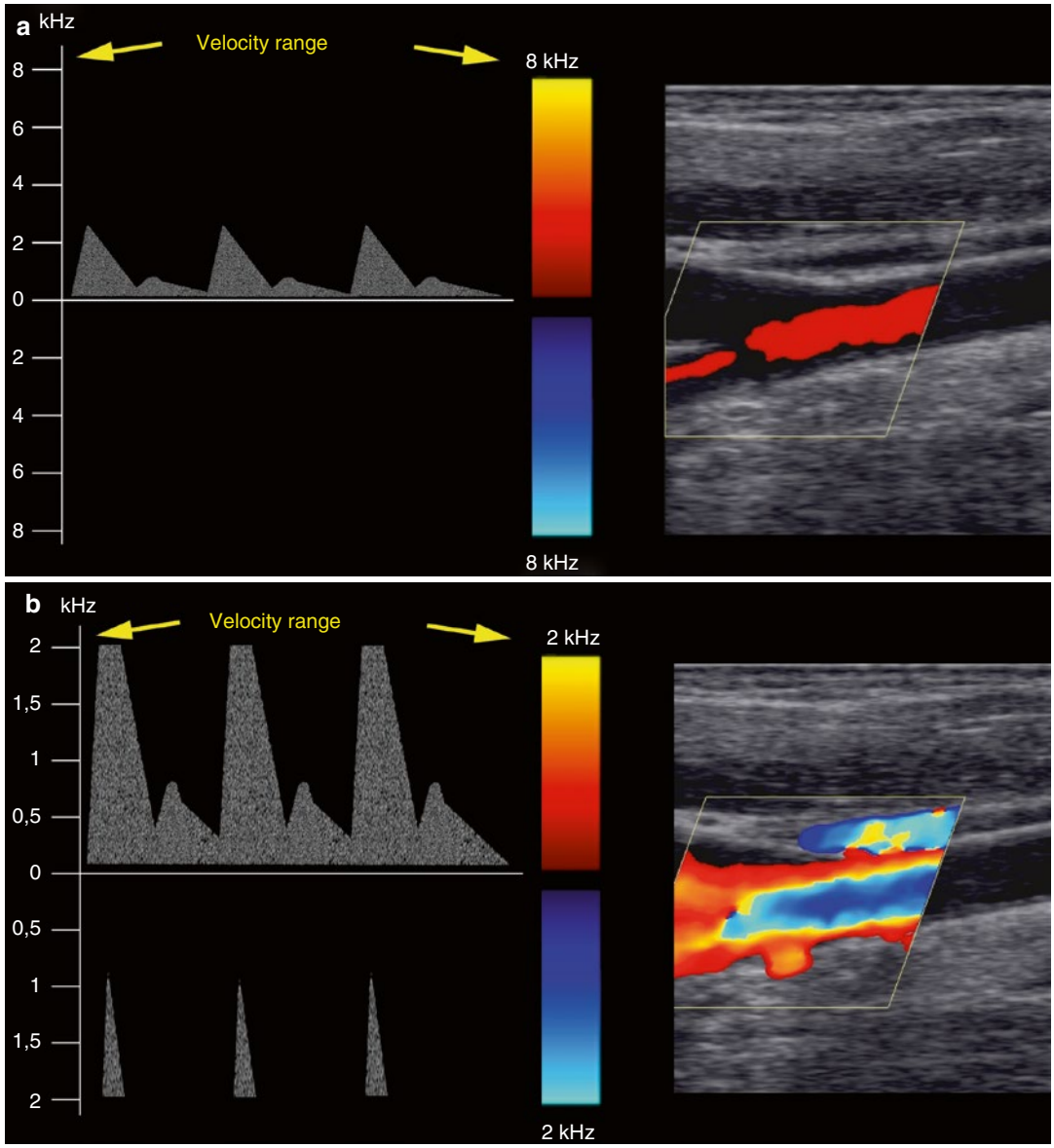

Fig. 1.26 Aliasing and color brightness dependence on the PRF

Every Doppler system allows the user to change the measurement range (and therefore the pulse repetition frequency) and adjust it to the prevailing flow rate.

If the measurement range is set too wide (see Fig. 1.26 left), optimum use cannot be made of the display area. The spectral curve of the conventional Doppler will be too small and too difficult to evaluate. On the colour-coded display, only a small part of the colour spectrum will be used, and the very slow flow velocities will not be visualised because the colour corresponding to the frequency will be too dark.

On the other hand, selecting a measurement range which is too small (low PRF) will mean that the Nyquist limit will be exceeded and the aliasing effect will appear (Fig. 1.26 right). In the Doppler spectrum, the peak speeds which occur during the systolic phase will be presented on the neighbouring channel, while, if the colour-coded Doppler is used, the higher speeds (often in the centre of a vessel) will be displayed in the wrong colour.

To determine the difference between the transmitted and received frequency generally, like the PW Doppler with a spectral display, the colour-coded Doppler needs several incoming signals one after the other (usually 5–15) in order to construct the Doppler frequency. This means that each Doppler scanning line (unlike the B-image lines) must be generated several times in quick succession at a frequency which corresponds to the PRF which has been set. One undesirable side effect of this unavoidable multiple scanning is the comparatively long time it takes to build up the colour-coded sample volume, so that the frame rate is significantly lower than that of the B-image. The examining physician is therefore well advised to overlay colour Doppler signals only on that part of the ultrasonic B-mode image which is necessary to show the vessel and to keep the so-called colour window as small as possible (Fig. 1.27).

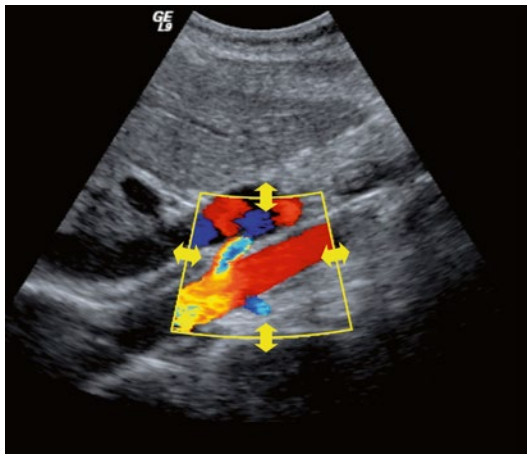

Fig. 1.27 Size adjustability of the color window

1.8 Triplex Mode

In Sect. 1.6, it was shown that duplex systems allow the PW sample volume to be positioned 'under view', while the virtual measurement gate assigned to the sample volume can be moved and placed within the B-image on the screen simultaneously. The same possibility exists if, instead of the conventional B-image, a colour Doppler image is used to visualise the target area and therefore to position the sample volume. Because three modes (B-image, colour-coded Doppler and PW Doppler with spectral display) operate in parallel in this case, this technique is called 'triplex mode'. The advantage of this is that the colour-coded Doppler is able to visualise the spatial differences in the haemodynamics, and this can be used as a navigation aid to place the sample volume accurately (Fig. 1.28). Thus, as well as providing a purely qualitative version of the colour-coded Doppler image, the triplex mode also allows the flow curves to be derived

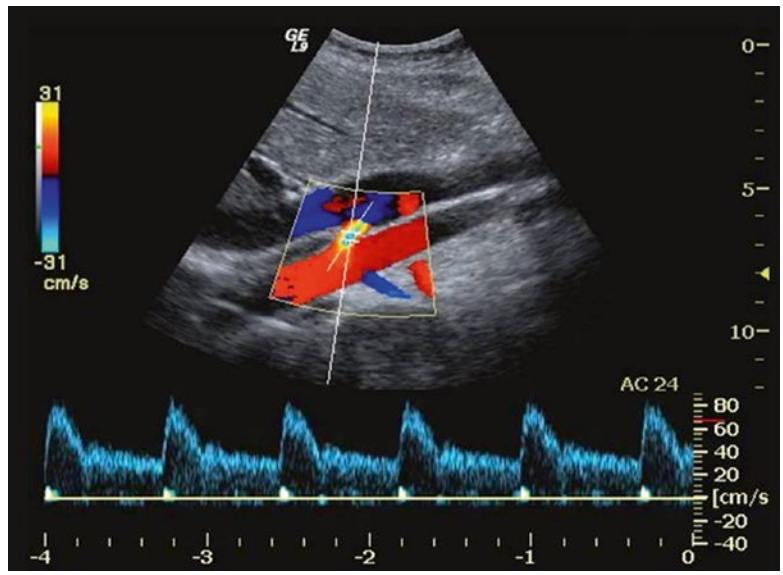

Fig. 1.28 Triplex mode – simultaneous display of the B-mode, color-coded Doppler signals and frequency spectrum

selectively in order to measure speeds and to evaluate the curve shapes semi-quantitatively.

Today's triplex systems allow the examining physician to choose between a genuine simultaneous real-time presentation of all the modes and an alternative mode where only the colour-coded Doppler image is generated during navigating, while the PW spectral Doppler is only active for deriving the flow curve and the colour-coded Doppler image switches to Freeze mode.

Pros and cons:

Real-time mode: Easier to use; the examining physician can control the true position of the sample volume at all times since displacements produced by movements can be immediately spotted and corrected.

Alternative mode: Better quality Doppler spectrum because no time resources are tied up to create the B-image and the colour lines and PW Doppler can operate at a higher pulse density.

1.9 Setting the Colour Threshold

Each colour sample volume has the approximate shape of a cylindrical body with diffuse edges and with a diameter and length ranging from several tenths to more than 1 mm. This size, which can hardly be reduced, is determined by the principle of operation and is primarily a function of the length of time of the transmission pulse packages necessary in Doppler mode. Since these spatial dimensions are so large, both static reflectors and flowing particles can occur in each sample volume simultaneously. In this case, the ultrasonic system would receive both B-mode echoes, which appear in grey, and coloured Doppler signals from the same location. It is for the examining physician to decide which of these two signals should be given the higher priority. A control element is available for this purpose. This is usually called a 'threshold' or 'balance', and it allows to prioritise the wanted signal by adjusting the threshold for grey scale signals. When a B-mode echo exceeds this level limit, the flow signals coming from the same location with a frequency shift are suppressed, and a grey shade is displayed on the screen which is proportional to the intensity. On the other hand, if the B-mode does not reach this level limit, they are discriminated against and the Doppler signals are given the higher priority. In order to visualise the level limit which has been set, a small mark is superimposed onto the reference greyscale bar (Fig. 1.29).

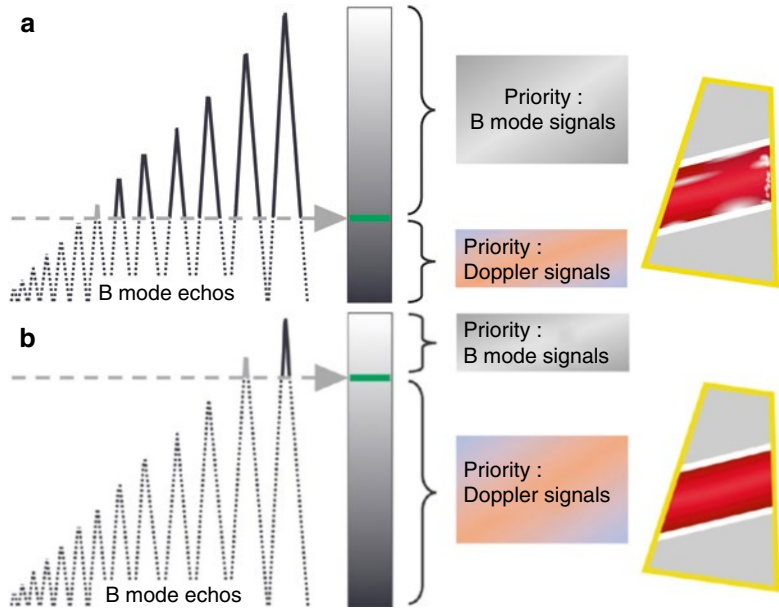

Fig. 1.29 Color threshold – setting of the priority if B-mode signals and Doppler signals occur at the same time in one sample volume

This B-mode signal is only displayed when the corresponding grey tone is above the mark. Grey shades beneath the mark, however, are only displayed when a Doppler signal is not present at the same time.

It is important to set the threshold correctly when the lumen of a vessel is interspersed with pronounced B-mode artefacts such as noise, side-lobe or reverberation artefacts. If the threshold is set too low (Fig. 1.29a), these artificial grey echoes will be prioritised and the flow signals will not be displayed filling the vessel. On the other hand, if the threshold is set too high, there is a risk that small structures giving weak echoes, such as fresh clots, will be masked by colour pixels and remain undiscovered (Fig. 1.29b).

1.10 Movement Disturbances and Wall Filters

When carrying out a Doppler examination on large arterial vessels, besides the frequency shift which comes from the flowing corpuscular blood components, high-intensity Doppler signals may also be produced by the pulsating vessel walls. In this case, the amplitude (not the frequency) of the echo can take on such large values that the receiver circuit of the Doppler system is overloaded and the flow curve superimposed by artefacts (Fig. 1.30). However, since vessel walls move at a lower speed than the blood components, it is possible to differentiate between the two signal sources according to their frequency.

Every Doppler system therefore has a so-called wall filter which discriminates against all Doppler signals beneath an adjustable frequency threshold. However, the wall pulsations of smaller vessels which are only weak usually do not require this filter to be activated. In fact, if the filter is switched on accidentally or the filter frequency is too high, the measurements may be full of errors which will have serious consequences. This can be clearly seen in Fig. 1.31. During the examination shown in the figure on the left, the wall filter was not switched on so the whole flow curve was displayed including the slow, end-diastolic velocities. During the automatic detection of systolic and diastolic frequencies for calculating the RI, therefore, it was possible to access all the necessary data and produce the correct value of around 0.7. From the curve on the right, it can be seen that the wall filter has been set to a very high level since there are no frequency points on the screen between this value and the zero line.

However, this means that the end-diastolic flow rates have been discriminated against and therefore do not appear on the screen.

When calculating the RI automatically or manually, it is assumed that the corresponding flow rate was almost 0, so the perfusion resistance of the circulation supply area must have been very high. In the example shown, therefore,

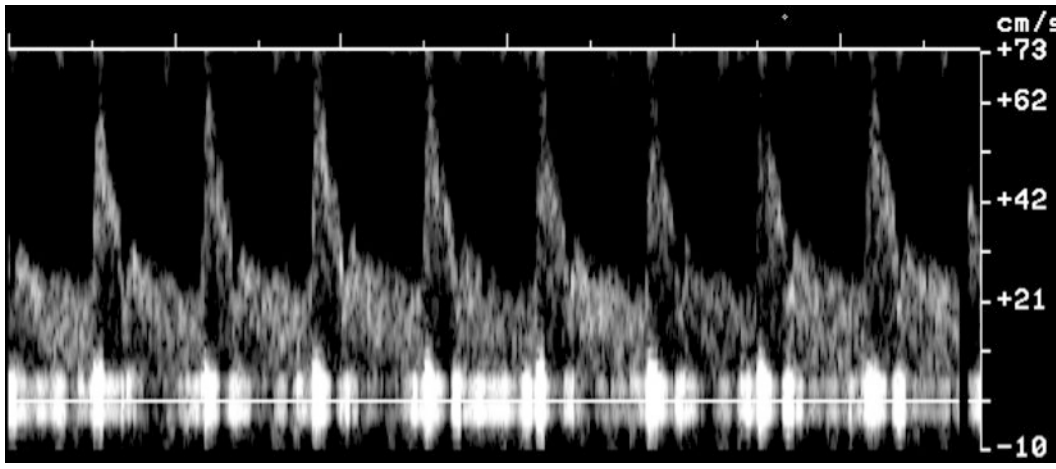

Fig. 1.30 Overlay of a Doppler spectrum through low frequencies caused by wall movements

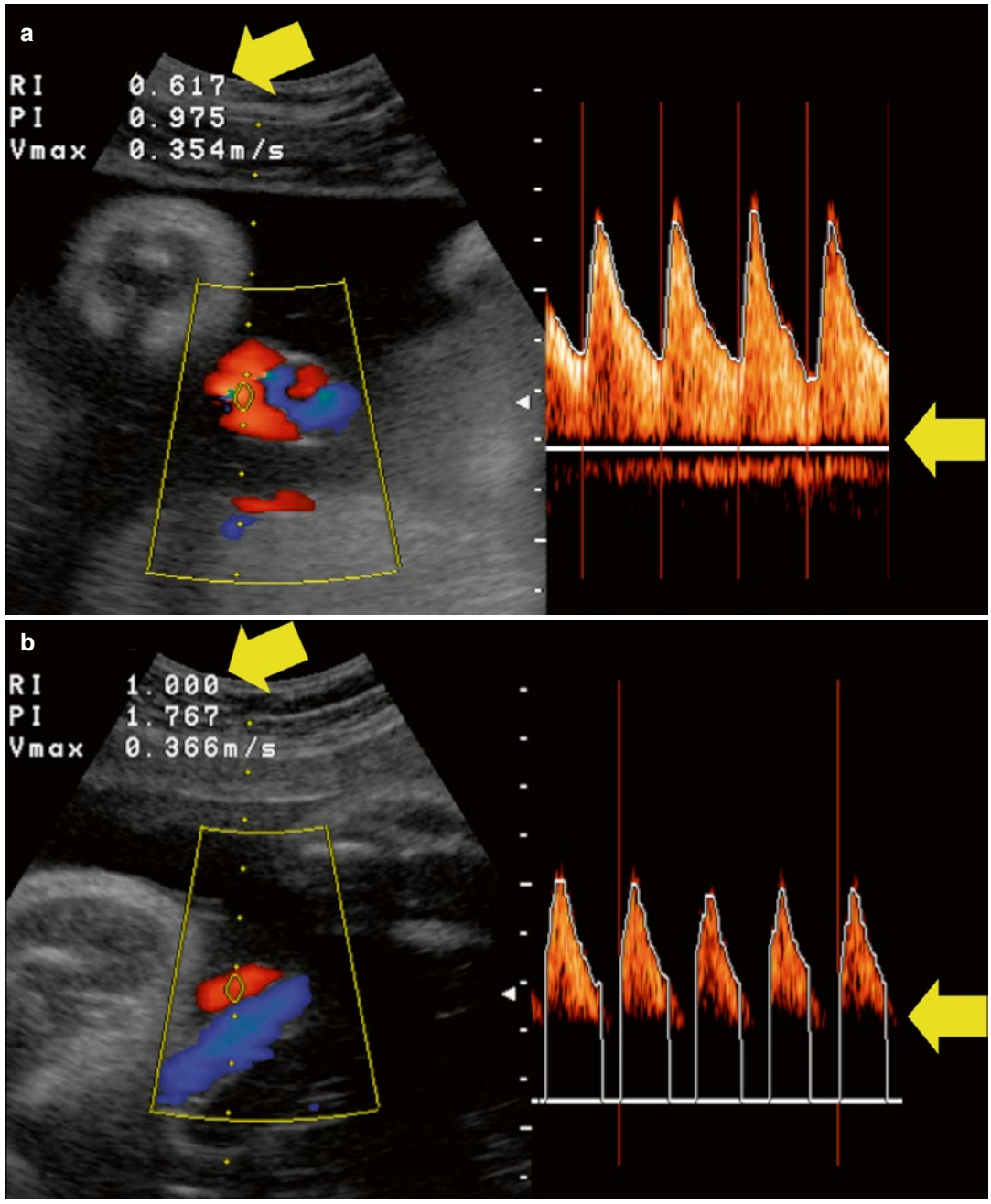

Fig. 1.31 Influence of the wall filter settings on the calculation of the RI

an RI of 1 was determined in error. One practical example of this is the assessment of the haemodynamic relevance of an open ductus arteriosus due to flow measurements on peripheral arteries.

Colour-coded Doppler systems are also fitted with filters for discriminating against low frequencies. Without these filters, soft tissues which are set in motion by respiratory excursions, arterial pulsations or other causes are displayed with large colour-coded areas which mask all the blood flows (so-called flash artefact Fig. 1.32).

Setting this filter in colour mode is somewhat delicate since, although a high filter frequency would exclude the flash artefact, the slow flow

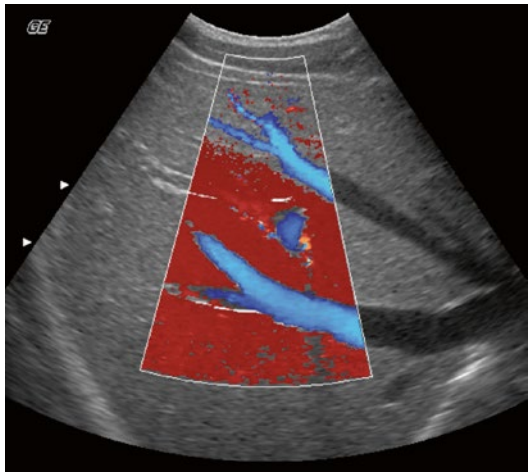

Fig. 1.32 Masking of vessels through flash artifacts

rates would still not appear on the screen. On the other hand, if the filter frequency is set too low, the flash artefacts would not be avoided so small vessels would not be recognised due to masking.

One practical example of this is the assessment of testicular blood flow in the case of suspected testicular torsion. In the low-flow region of the testicle, the vascularisation can only be presented when a low wall filter is selected. If the wall filter used is too high, the blood flow will not be detected even if present, thus resulting in a false diagnosis of testicular torsion.

1.11 Adjusting the Gain

The reflections arriving at the piezoelectric transducers of an ultrasonic probe induce a tiny electrical voltage of only a few microvolts. This low voltage must first be amplified of course for further signal processing, particularly for converting it into digital signals.

However, since reflections show a wide variety of amplitudes depending on where they come from, the amplifier must have a gain control so that the signals can be adjusted to suit the operating range of the equipment.

Each PW-Doppler or colour Doppler module must have an appropriate gain control as well as the signal processor for the 2D image.

When using the conventional spectral Doppler, the amplifier must be set so that maximum frequency points (i.e. the ends of the envelope) can be clearly seen and clearly stand out against the background over the entire curve. The gain should be adjusted so that it can be seen that the different frequency dots on the spectrum differ in brightness. If the gain is too high, we will not be able to see any differences in brightness and we will not be able to detect any graduations grey. Furthermore, the background, i.e. the area outside of the spectrum, should be free of noise if possible.

The correct adjustment of the gain according to the recommendations described above is of fundamental importance for the automated, semi-quantitative analysis of frequency spectra. An envelope end which cannot be clearly identified by the automatic system will always produce errors when calculating the indexes.

During an examination using the colour-coded Doppler, the gain must be constantly adjusted to the different signal intensities which vary considerably.

If the amplification is too weak, the vessel which is under examination either remains completely echo-free (therefore black) or is only partially colour coded (Fig. 1.33 left).

On the other hand, increasing the gain too much will also lead to an undesirable increase in spurious signals. These appear on the colour-coded image as ‘colour noise’ – as diffuse, rapidly changing, flashing pixels of short duration (Fig. 1.33 right). In this case, the amplification would of course have to be reduced so that the smallest flows are not masked by the colour noise.

1.12 Power Doppler

Most of the usual colour-coded Doppler systems today offer a special mode for visualising particularly low-flow velocities in the smallest vessels. This mode, which is called Power Doppler, Angio or Perfusion mode, is used to visualise the perfusion of an organ qualitatively, especially a transplant, or even to evaluate the perfusion of a tumour (Rubin et al. 1994).

In Power mode, the proportionality between velocity and colour intensity which otherwise occurs with colour-coded Dopplers is cancelled.

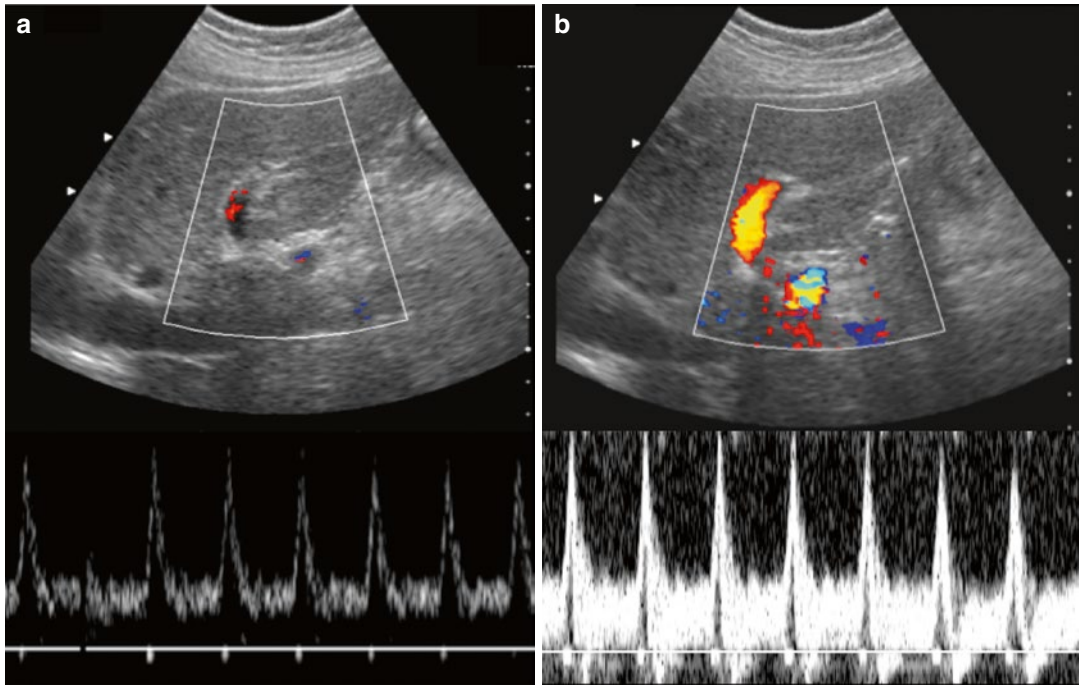

Fig. 1.33 Effect of gain control

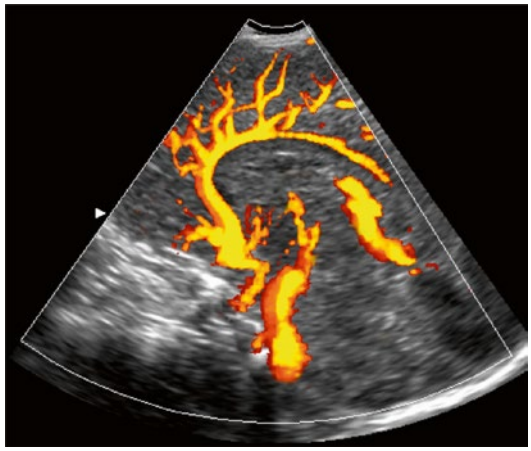

Fig. 1.34 Power Doppler, no proportionality between velocity and color brightness

Regardless of how fast the reflecting corpuscles are moving, all flow velocities are presented in the same colour. Only the reflection intensity has an effect on the colour brightness – the more reflecting corpuscles present in the sample volume, the brighter the corresponding area is on the screen.

Figure 1.34 shows the flows in a lymph node recorded in Power mode. In this case, the flow signals were not shown in red but in green, and the echo signals of the tissue were eliminated.

1.13 3D Dopplers

In many areas of application for sonography, three-dimensional scanning is now established as an additional mode alongside the conventional two-dimensional section image method (Riccabona et al. 2003; Raine-Fenning et al. 2004).

In this case, the array:

- Is either guided manually and therefore comparatively slowly over the region being examined (Static 3D) or oscillates, motor driven, in a housing filled with fluid and scans the anatomical structures quickly and periodically (4D sonography).
- The latest technique which can be included is a matrix array with a two-dimensional (usually square) arrangement of numerous tiny crystals for volumetric scanning. These ultrasonic probes allow the sound field to be swung quickly without mechanically moved parts in order to provide a three-dimensional scan of

structures which also pulsate, such as the heart.

- Regardless of the technical implementation, with all 3D methods, the successively acquired section images are correctly lined up together in terms of location and angle while scanning is still in progress, so that a three-dimensional echo block is available immediately after the sweep has finished (Fig. 1.35). This block resembles a cube or more like a truncated pyramid depending on the scanning method.

The special diagnostic value of this 'Volumetric Echo Recording' is that it is possible to use the echo block to reconstruct any virtual sectional plane. Thus, planes of observation can be produced which would never have been possible using 2D sonography due to the relationship between the position of the ultrasonic probe and the region under examination, e.g. sections parallel to the surface. As shown in Fig. 1.36, the type of section extracted can also contain flow information recorded by Doppler sonography.

As well as the option of extracting any section from the volumetric block, 3D systems also offer the examining physician quite different modes of presentation, such as 'transparency mode', a method also known as 'glass-body rendering'.

This mode features plastic presentation of the entire recorded echo block, where the external surfaces facing the observer and all the other sections are displayed partially transparent rather than solid. Each section image therefore enables the examining physician to see the echo structure of the level behind so the entire box is equivalent to a so-called glass body.

If this glass-body rendering is combined with Colour or Power Doppler mode, then vessel structures with their numerous branches can be displayed concisely within their spatial context. The echoes of the soft structure can be suppressed so that, as with the angiography, only the vessels which contain flowing blood appear on the screen (Fig. 1.37).

1.14 Non-Doppler-Based Flow Recording

The physical and technical limitations of the Doppler sonographic method for detecting flows have recently led to intensification in the search for an alternative technology. Although this kind of alternative to the PW and CW Doppler with spectral presentation has not been found so far, one favourable substitute for colour Doppler technique (which is particularly susceptible to artefacts) has recently become available. The new method is based on a special analysis of the

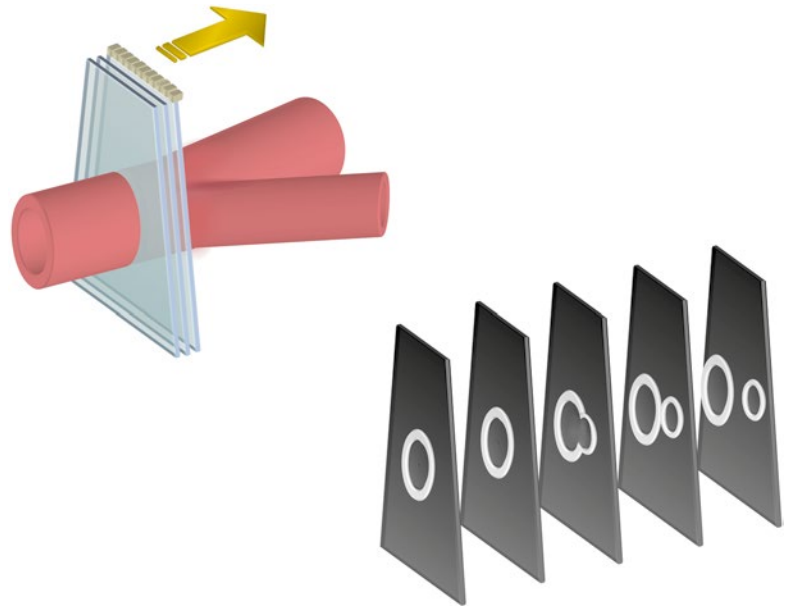

Fig. 1.35 Scan of a volume by generation of multiple parallel 2D planes

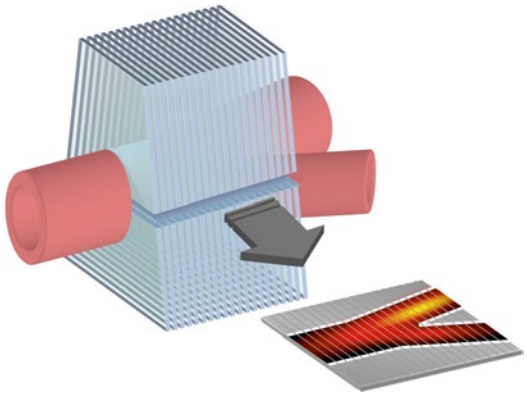

Fig. 1.36 Reconstruction of a plane parallel to the surface (C-plane)

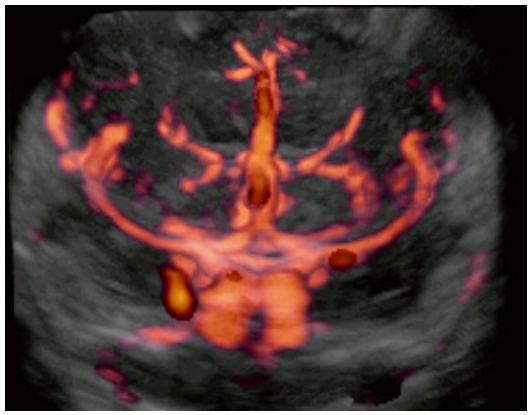

Fig. 1.37 Glass-body rendering of intracranial vessels

B-image and is therefore also known as 'B-flow' (Chiao et al. 2000).

In order to detect the movement, each B-image scanning line is constructed twice within a short space of time and the signals received compared with one another. Since it is essentially only the corpuscular blood components which relocate locally during this short cycle of approximately 1 ms and not the tissue, the latter is subtracted from the display, so that the reflectors in the flowing blood clearly dominate in relation to their environment (Fig. 1.38). The faster the blood flows, the greater is the difference between the two scans and the brighter the corresponding light spots are on the screen. The new technique therefore corresponds to a subtraction procedure but where the proportionality between flow velocities and brightness is only approximately linear. By adding the detection of frequencies altered by the Doppler effect, the direction of the flow can also be found and visualised by colouring the B-flow signals.

B-flow offers a substantially higher spatial resolution (no overwriting by the vessel walls) compared to colour-coded Doppler sonography because the short transmission signals of the B-mode are used. Vessels which are close in proximity can be distinguished from one another much more easily (Fig. 1.39) – stenosis and pathological

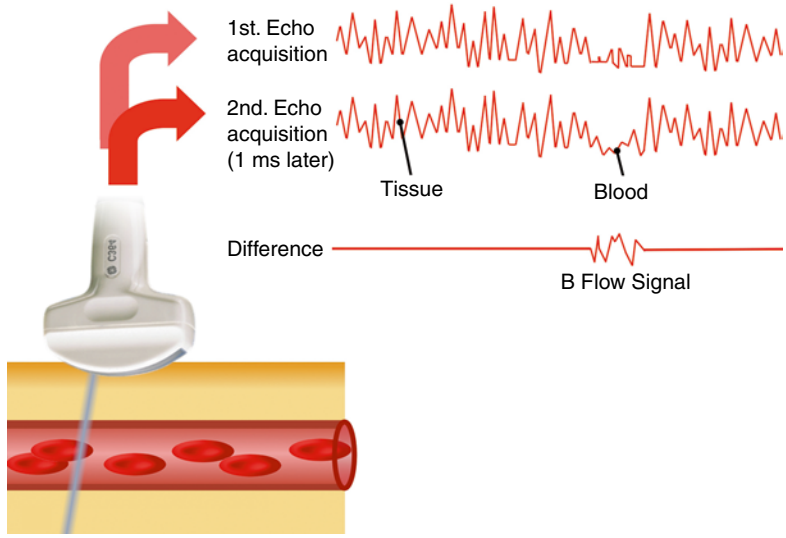

Fig. 1.38 B-flow – capturing of flow signals according to the subtraction method

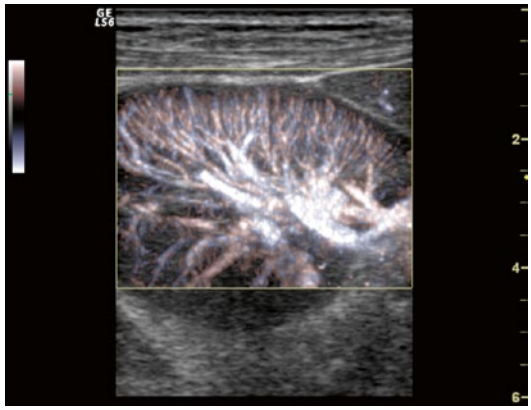

Fig. 1.39 High spatial resolution B-flow. Image of a renal perfusion

vessel conditions are much clearer to see and easier to assess because of the detailed nature of the presentation (Wachsberg 2007). Studies on vessels narrowed by arteriosclerosis have shown that plaque contours are accurately visualised by means of B-flow and allow valid measurement of the remaining lumen (Weskott 2000). The new technique also offers a significantly higher frame rate since, unlike colour Doppler, it is not necessary to generate each scanning line 5–15 times one after the other. As a consequence of this higher frame rate, the haemodynamics on the display is much closer to reality since rapid changes in flow and direction, such as short regurgitations, are reliably displayed.

Another advantage of the new mode in comparison to Doppler sonography is its independence to the scanning angle. Even at an angle of 90° between the flow axis and the ultrasonic beam, the signal strength largely corresponds to a signal that would be obtained from an insertion angle of 0°.

One disadvantage at the present time is still the somewhat limited depth of penetration in comparison to the conventional colour Doppler. With technical progress, however, these limitations may be overcome in the foreseeable future so that B-flow in combination with the CW/PW spectral Doppler may completely replace the triplex process usually used today.

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