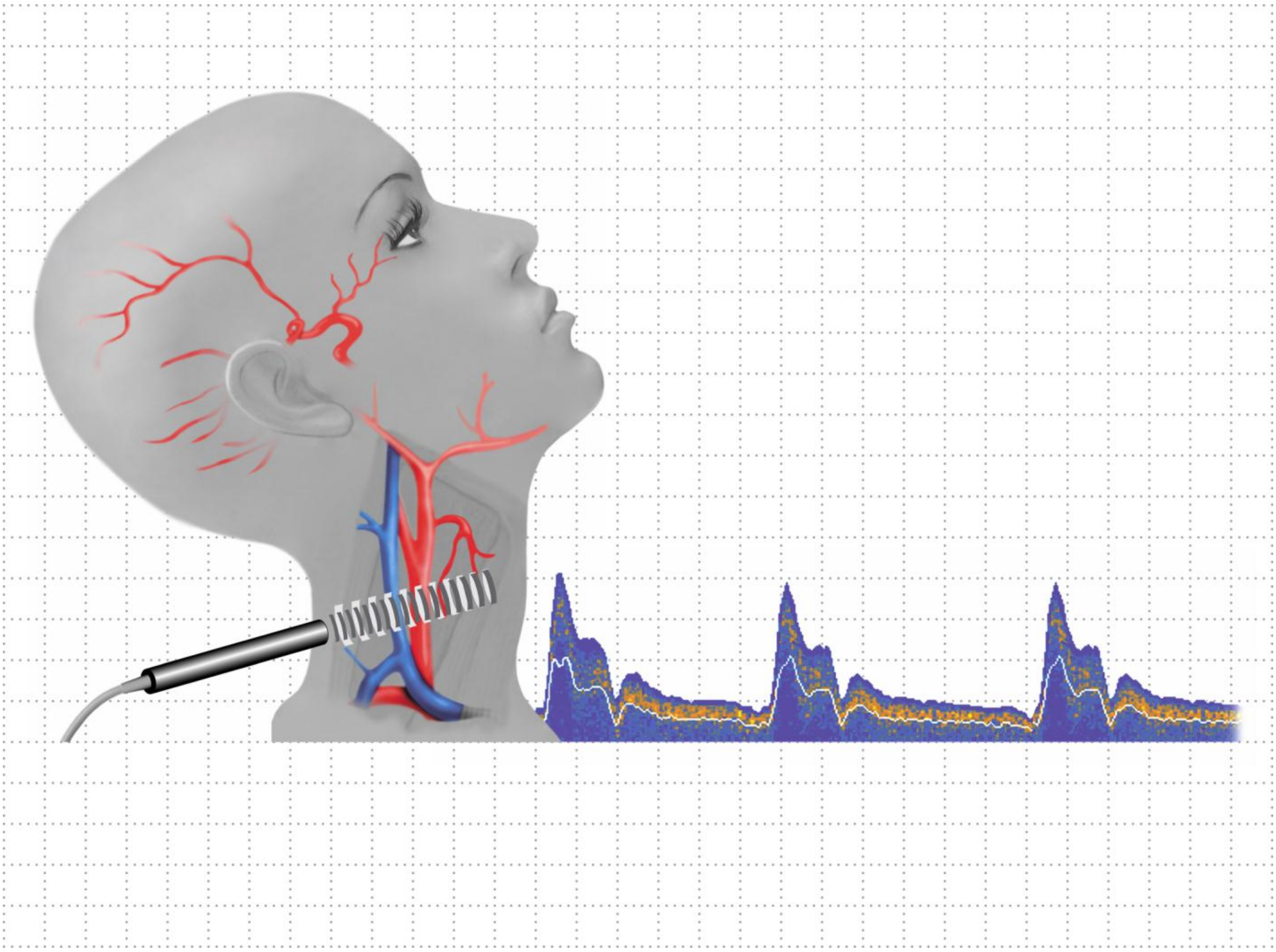
Doppler sonography in vascular diagnosis



ELCAT GmbH
innovative medical products

Doppler Sonography in Vascular diagnosis

Preface

Amongst angiologists, the great importance of Doppler sonography in vascular diagnosis has been undisputed for many years. For example, in his preface to a collection of papers published by Thieme in 1978, Bollinger/Zurich wrote: “Ultrasound Doppler technology represents the greatest enhancement in angiological diagnostics in the last 10 years“. In the preface to his book “Praktische Dopplersonographie (Practical Doppler sonography)“ published in 1984, M. Marshall/Munich demanded .”…that this method should be adopted as widely as possible in practice“.

In the meantime, ultrasound Doppler examinations are part of routine examinations in angiology in daily practice. Angiologists have a wide range of Doppler equipment with different technology for a wide range of applications available to them.

This brochure is meant to provide an overview of the different equipment and their application possibilities – from unidirectional and bidirectional Continuous-Wave Doppler devices via frequency analysis all the way to pulsed Doppler for transcranial Doppler sonography. In addition, it also deals with the physical principles of sound and the Doppler effect and provides a brief insight into duplex and colour duplex sonography.

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The Nature of Sound 1.1.

Sound waves – or mechanical waves in general – are usually generated by an oscillating body, for example the resonance bow of a string instrument or the oscillating prongs of a tuning fork. As shown in Fig. 1.1.1, during an outward oscillation, the oscillating prongs of the tuning fork briefly compress the molecules in their immediate vicinity. A defined area of increased molecule density is created (excess pressure zone). This excess pressure zone spreads in the direction of the oscillation in a kind of chain reaction.

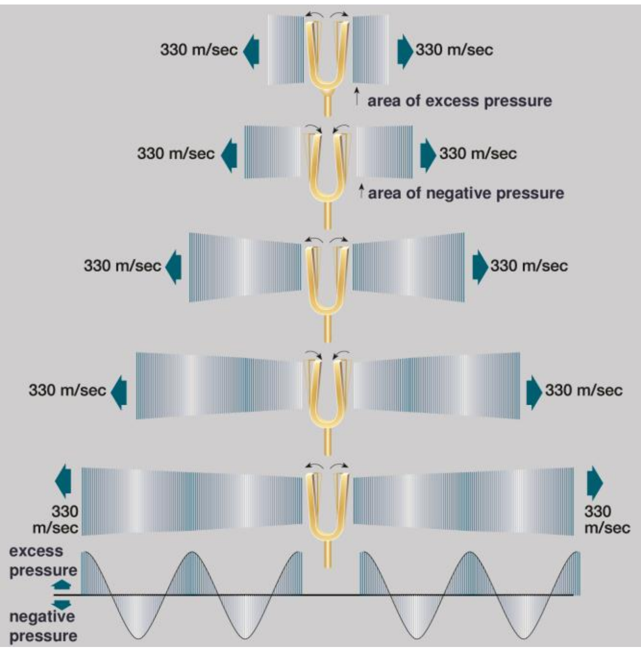

Fig. 1.1.1
Generation of a sound wave by a tuning fork in the form of cyclical pressure differences

Whilst this area of excess pressure is spreading, the prongs of the tuning fork move inwards again. This creates a slight suction which forces the molecules to follow the movement of the prongs. As a result the molecule bonds are stretched somewhat. This in turn creates an area of reduced molecule density (area of negative pressure). This area also spreads within the medium and follows the area of excess pressure created before.

These pressure areas spread due to the elastic bonds between the molecules. The molecules do not move through the media, but merely oscillate around their fixed resting point (Fig. 1.1.2).

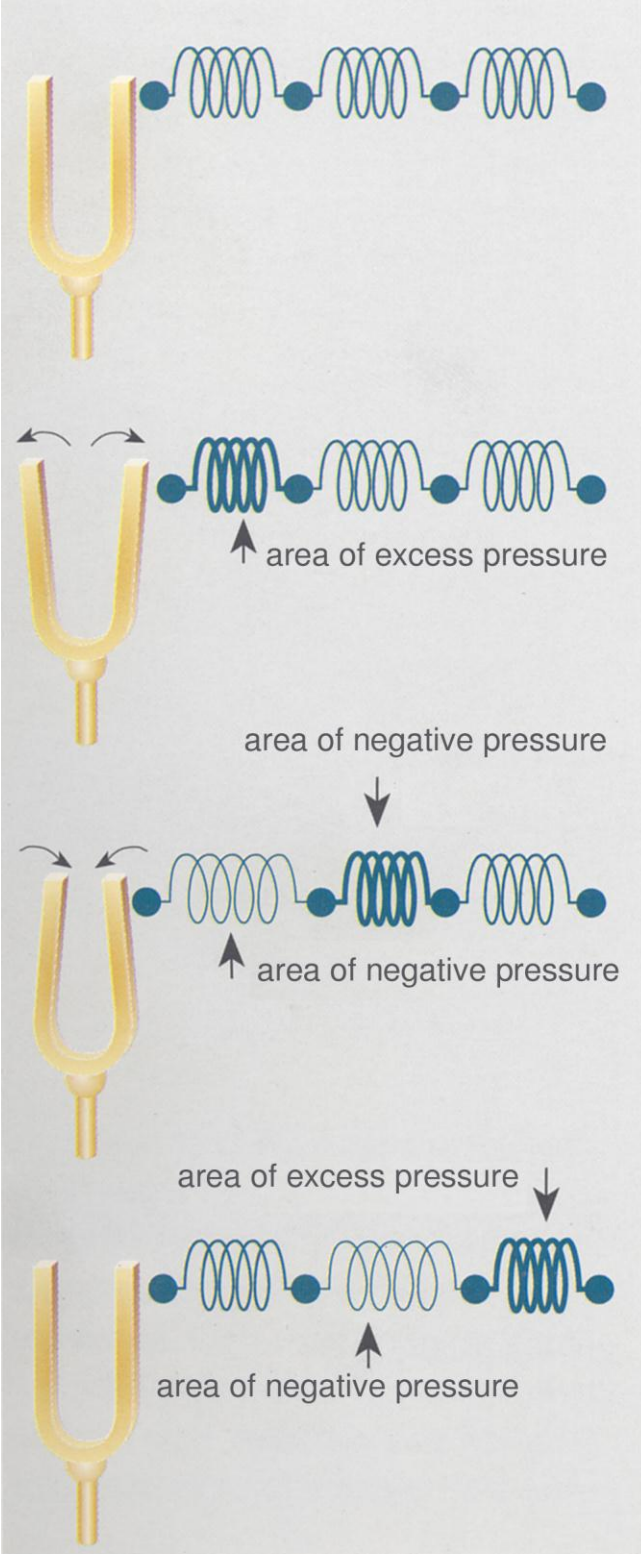

Fig. 1.1.2
Idealized representation of the spreading of cyclical pressure differences based on the elastic connection (spring) of the molecules

The speed at which waves spread or the speed of sound depends on the elasticity and molecule density of the medium. Here are some examples:

Air	330		m/s
Water	1.480		m/s
Soft tissue	approx.	1.540	m/s
Bone	2.700	– 4.100	m/s

Sound waves - created by oscillating bodies – spread in the form of cyclical pressure differences and the speed at which they spread is material specific.

If a body is oscillating quite slowly, the distance between the pressure zones, the so-called wave length, is large. The human ear picks up this oscillation as a deep note. In contrast, a body which is oscillating quickly generates a short wavelength and therefore a high note.

For example, the soundboard of a double bass only oscillates approx. 30 times a second when the deepest note is played. It generates sound waves with a wavelength of approx. 10 m.

If the highest note on a violin, the 4 accented b is played, the soundboard oscillates approx. 4,000 times a second thus generating a wavelength of approximately 8 cm (Fig. 1.1.3).

The pitch also depends on the number of pressure zones arriving at the eardrum and therefore also on the number of oscillations per unit of time, the frequency.

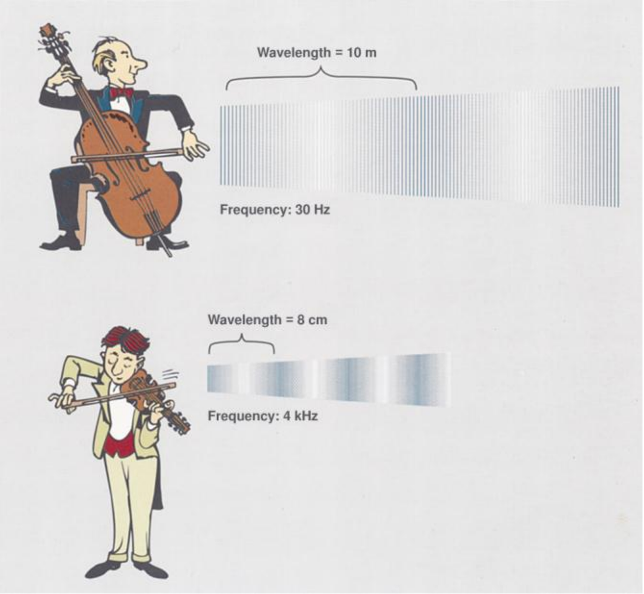

Fig.1.1.3
Generating different wavelengths depending on the oscillation frequency

The oscillating frequency is defined in “Hertz“ (Hz). 1 Hz describes 1 oscillation per second. 1,000 Hz is described as 1 kilohertz (1 kHz) and 1,000,000 Hz is defined as 1 Megahertz (1 MHz).

The human ear can hear a frequency range between approximately 16 to – depending on age – approx. 10,000 to 20,000 Hz. Frequencies outside this range and below 16 Hz are described as infrasound, frequencies above 20,000 Hz are described as ultrasound.

Generating Ultrasound 1.2.

Thin slices made of quartz crystal are used to generate the extremely high frequencies between 1 and 10 MHz used for medical diagnostic purposes (Fig. 1.2.1).

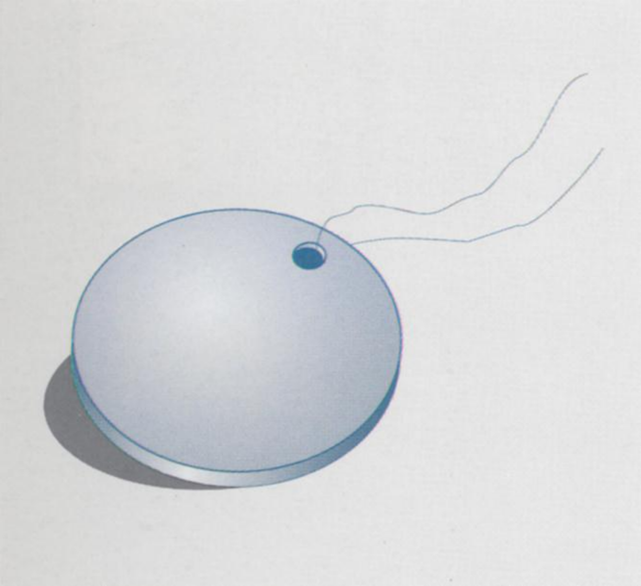

Fig. 1.2.1
Quartz crystal slice with conducting wires

These quartz crystal slices are coated with a thin layer of metal on both sides and one conducting wire is soldered to each side. If a direct voltage is applied to these wires, an electrical field is created between the two metal layers.

It is interesting to observe, that due to this electrical field, the atomic charges of the crystals repel each other. The crystal grid therefore occupies a slightly larger space - the crystal expands slightly. If the polarity is reversed now – i.e. if plus and minus are reversed – the charges attract each other and the crystal grid occupies a little less space - the crystal becomes a little thinner. (Fig. 1.2.2).

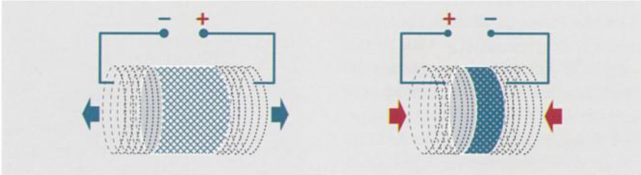

Fig. 1.2.2.
Changes in spatial expansion of a quartz crystal grid by changing voltage polarity

The mechanical movements of the crystal follow the voltage polarity virtually with no inertia. If a constantly reversing polarity is generated using an alternating current generator, the crystal changes size in the rhythm of the alternating voltage. The crystal therefore is turned into an oscillating body and generates sound waves in accordance with the frequency of the alternating voltage applied (Fig. 1.2.3).

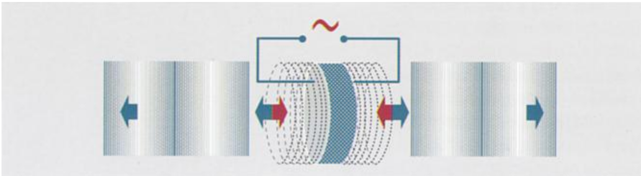

Fig. 1.2.3
Generation of sound waves using a quartz crystal and applied alternating voltage

However, quartz crystals are not only suitable for radiating but also for receiving the ultrasound echo returned by the tissue. If the crystal is hit by mechanical sound waves, i.e. areas of excess pressure and areas of negative pressure, it in turn generates an electrical alternating voltage with the frequency of the oscillation received. (Fig. 1.2.4). This phenomenon is known as the piezo effect and was discovered in 1880 by Pierre Curie.

As the amplitude of the voltage received is very small, it is amplified using an HF-amplifier (high frequency amplifier and then transmitted to the electronic system of the equipment.

This transmission and reception technique of ultrasound with the help of a quartz crystal is found in all diagnostic medical ultrasound systems – whether in Doppler or image equipment, whereby the form of the crystal can differ.

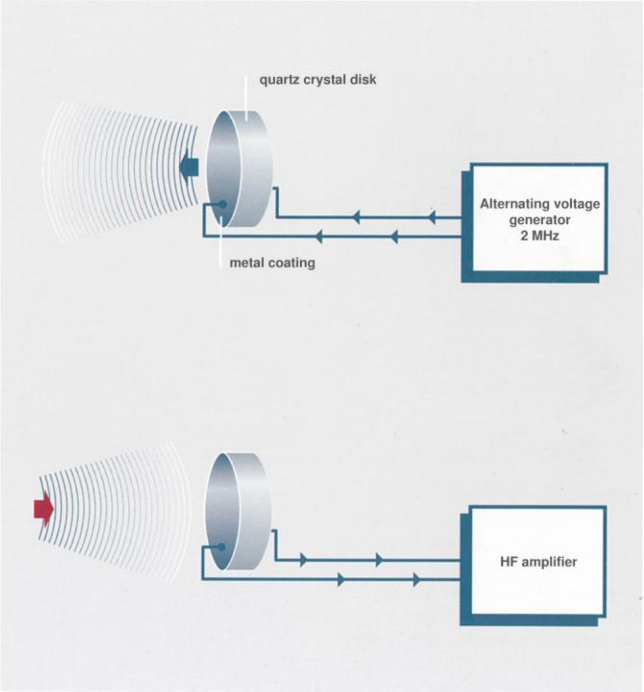

Fig. 1.2.4
Transmission and reception technology of ultrasound using a quartz crystal (piezo effect)

The Doppler Effect 1.3.

The Doppler effect is named after the Austrian mathematician and physicist Christian Johann Doppler (1805-1853). Doppler had observed that the light spectrum of a celestial body which is approaching the observer was displaced to the blue end of the spectrum, i.e. it reached the Earth with a short wavelength, whereas the light spectrum of the same celestial body which was now moving away from the observer was displaced to the red end of the spectrum and therefore arrived at the Earth with a longer wavelength (Fig. 1.3.1). Based on this observation, in 1842 in his work “On the coloured light of double stars“, he assumed that an observer receives a different frequency than the actual frequency as soon as the transmitter and / or recipient move relative to one another. If the transmitter and he recipient move towards one another, the frequency received by the recipient is higher than that transmitted by the transmitter. And vice versa, the frequency received deviates from the transmitted frequency towards the bottom end of the spectrum as soon as the transmitter and recipient move away from one another.

However, this phenomenon, which is very difficult to demonstrate using light, when applied to the acoustics relevant to Doppler sonography, soon becomes clear using the example of the tuning fork.

As we have already described, the pitch of a note depends on the wavelength of the sound. The larger the number of areas of excess pressure arrive at the ear of the observer per second, i.e. the shorter the wavelength, the higher the pitch of the note we hear.

Now, if we observe the tuning fork shown in Fig. 1.3.2, it oscillates 1,000 times per second and produces a sound field with a wavelength of approx. 33 cm. If one observer was placed on each side of the tuning fork, each observer would be conscious of a frequency of 1,000 Hz.

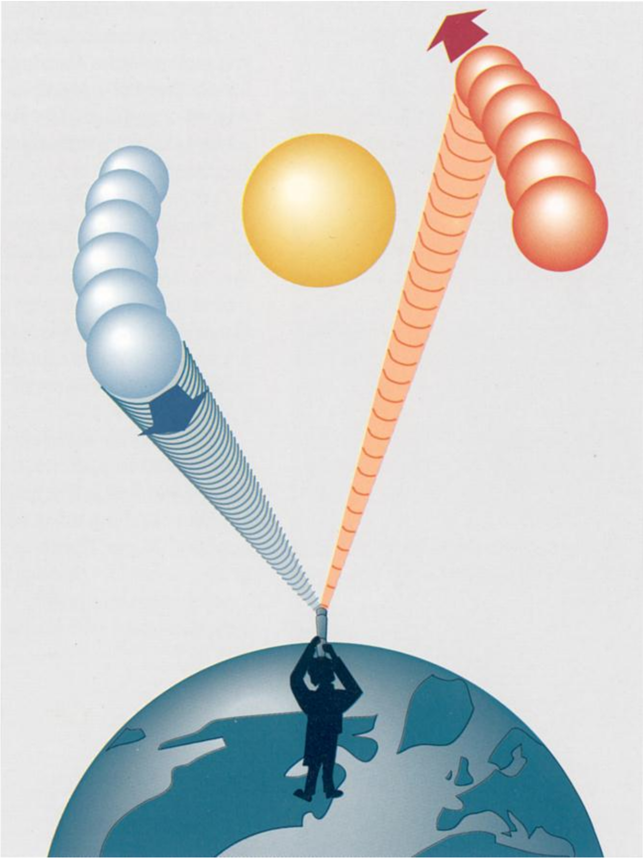

Fig. 1.3.1
Displacement of the light spectrum of a star from red to blue depending on the direction of movement relative to the observer

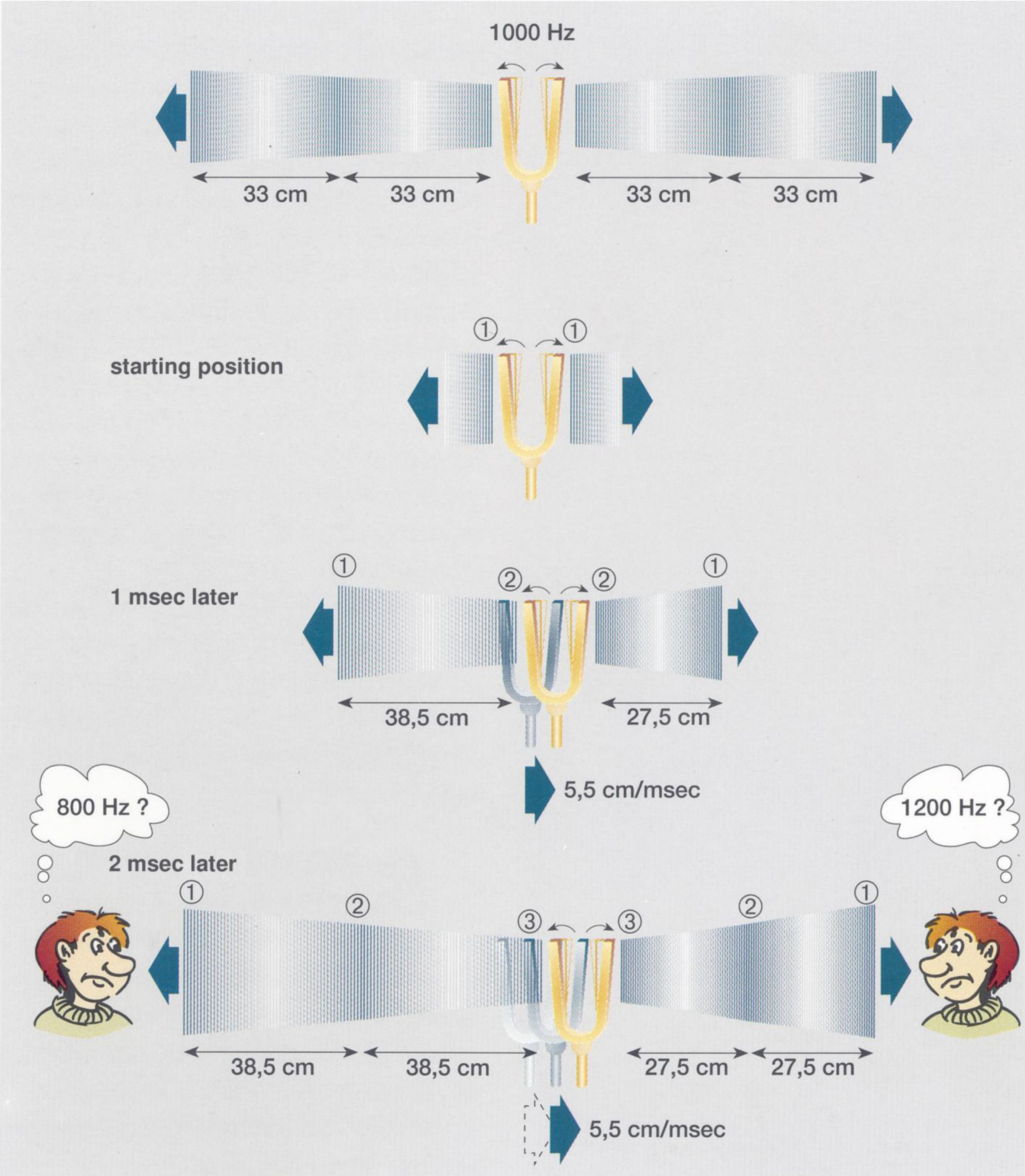

Fig. 1.3.2
Displacement of the frequency due to a decrease or increase in wavelength due the movement of the transmitter itself. Depending on the direction of movement relative to the recipient, the displacement in frequency is negative or positive

The next example shows the starting position of the tuning fork which is moving to the right at a speed of 200 km/h. The prongs of the tuning fork are oscillating to the outside and generating the excess pressure zone labelled with “1” and which is now spreading through the medium at the speed of sound.

One millisecond later the prongs of the tuning fork are swinging outwards again and are generating the area of excess pressure labelled with “2“. Due to its inherent speed the tuning fork has moved 5.5 cm with the same period of 1 millisecond. This means that the 2nd area of pressure was generated 5.5 cm to the right relative to the starting position.

Therefore the wavelength on the right hand side is decreased by 5.5 cm to only 27.5 cm. In contrast, the wavelength on the left hand side increases by 5.5 cm to 38.5 cm. If the tuning fork stays at the same speed, a sound field is generated on the right hand side, the wavelength of which is only 27.5 cm, whereas a sound field is created on the left hand side, the wavelength of which is 38.5 cm. Our observer on the right hand side would now detect a frequency of approx. 1,200 Hz, whereas the observer on the left hand side would only detect a frequency of approximately 800 Hz, **even though the tuning fork continues to oscillate at a frequency of 1,000 Hz.**

Therefore the receiving frequency increases relatively to the transmission frequency if the source of sound moves towards a stationary observer. In contrast, the frequency received decreases if the source of sound moves away from a stationary observer.

The Doppler effect also occurs in the reverse situation i.e. if the source of the sound remains stationary but the recipient moves relatively to the source of the sound.

In Fig. 1.3.3 the observer moves towards the sound waves coming towards him. In contrast to the stationary observer, he therefore also receives pressure zones which occur along a distance of 5 m every second. Using the frequency of 440 Hz shown, this would be equivalent to an additional 7 pressure zones. Instead of the transmitted frequency of 440 Hz, the frequency received would be 447 Hz. If the observer were to move in the direction in which the sound is spreading, i.e. with the sound wave, his ear would be hit by pressure zones less frequently per second than the ear of the stationary observer. Therefore he hears a deeper note of 433 Hz (Fig. 1.3.4).

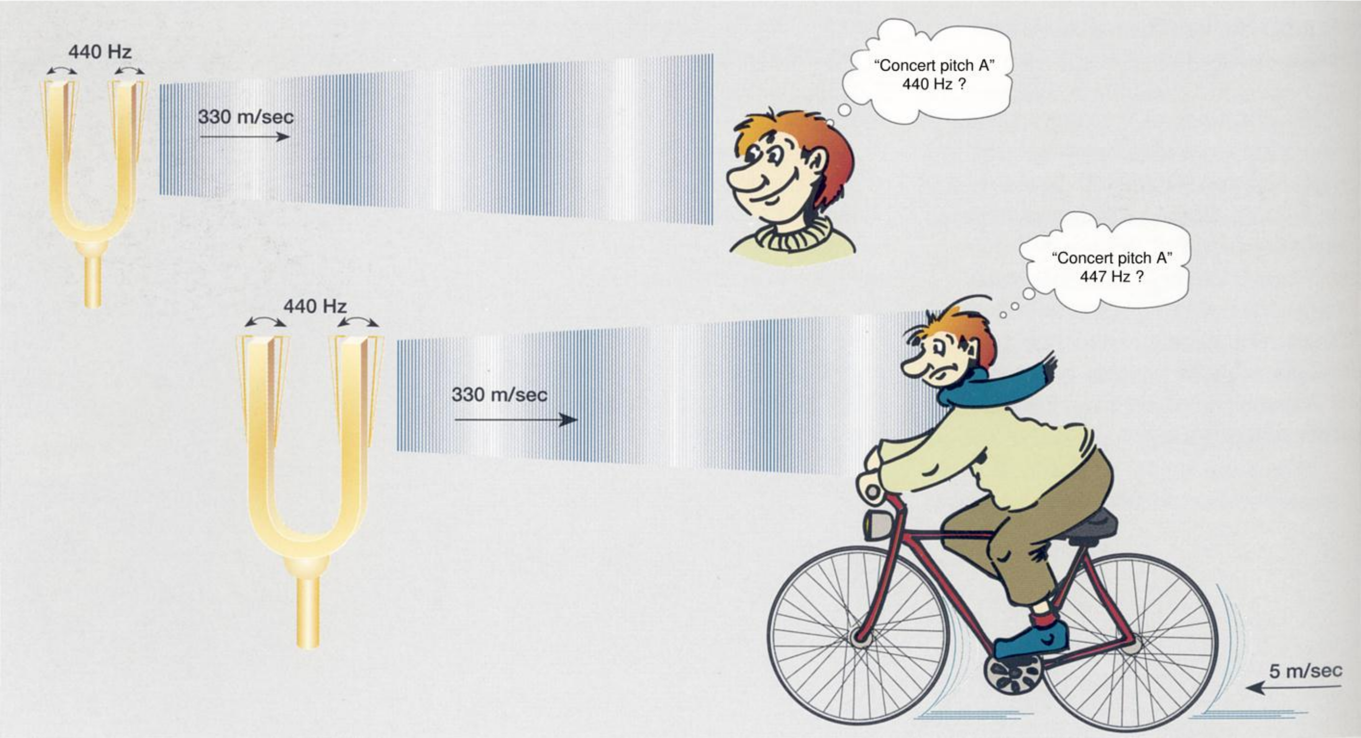

Fig. 1.3.3
Positive frequency displacement when the recipient moves towards the transmitter in relation to the sound wave

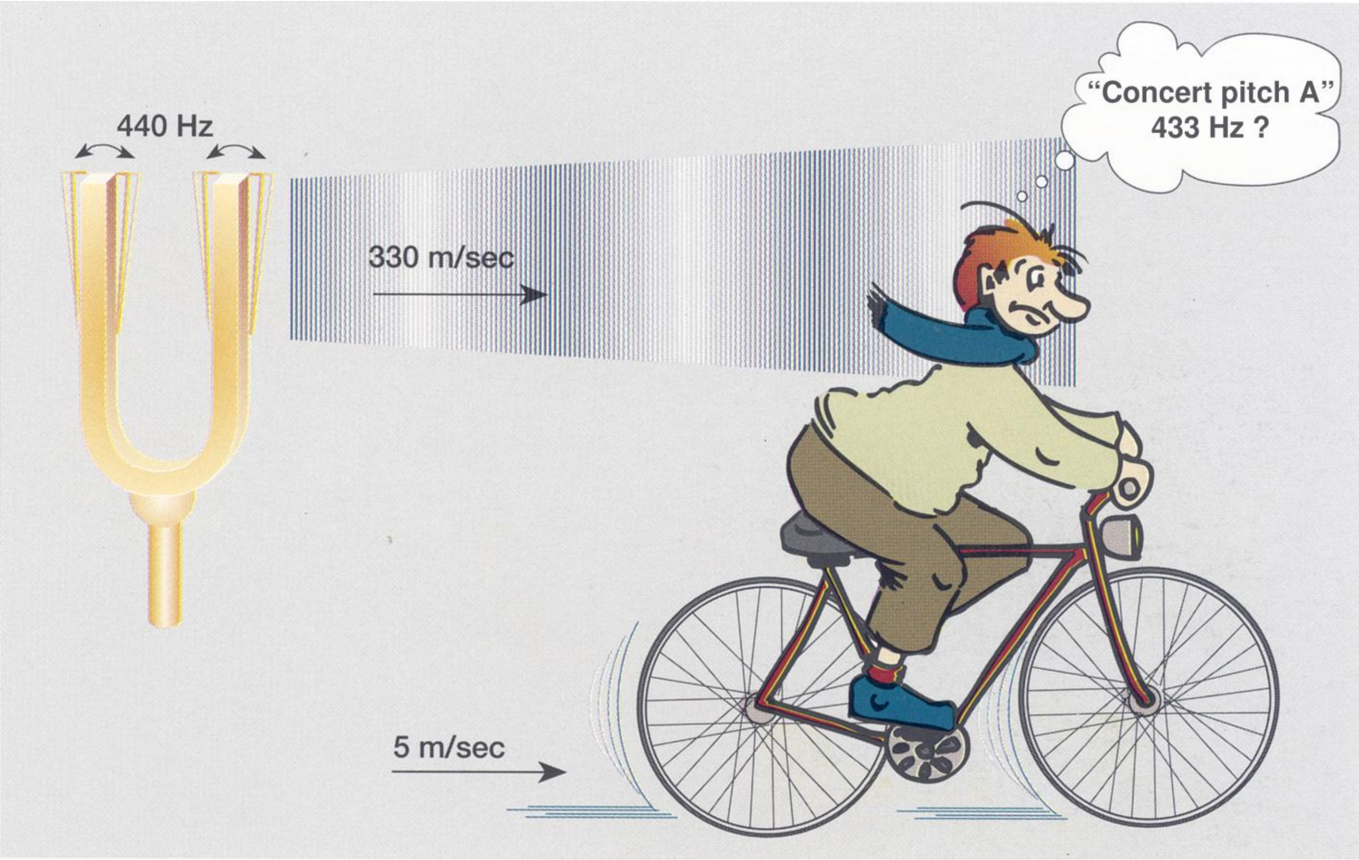

Fig. 1.3.4
Negative frequency displacement when the recipient moves away from the transmitter.

Even though it is not irrelevant whether the transmitter or recipient are moving towards or away from each other when calculating the frequency displacement, a simplified equation can be specified due to the relatively low speed of the recipient or transmitter in relation to the speed of sound:

$$F_d \approx \frac{F_o \cdot V}{C}$$

Here:
F_d = Frequency displacement in Hz
F_o = Transmission frequency in Hz
V = Velocity in m/sec
C = Speed of sound in m/sec

Therefore F_d is proportional to the transmission frequency F_o as well as velocity V of the transmitter or recipient and is inversely proportional to the speed of sound C within the medium.

As unknown as the Doppler effect seems to be, it accompanies us through our daily lives. For example, you are standing at a busy road or racing track and a vehicle passes you at constant speed, the sound you hear when the vehicle is approaching you seems higher than when the vehicle is moving away from you even though the engine is in actual fact producing the same sound all the time (Fig. 1.3.5).

Fig. 1.3.5
Doppler effect using a passing racing car as an example

In a last case, the Doppler effect also occurs if the recipient and transmitter are stationary but the sound transmitted is deflected by a moving reflector. In this case, a double Doppler displacement occurs. The first displacement occurs when the sound hits the reflector and a second displacement occurs when this sound is reflected at an already increased frequency (Fig. 1.3.6).

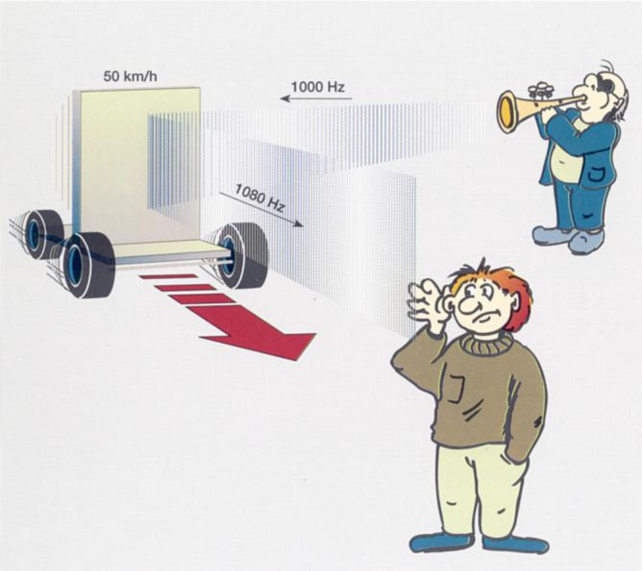

Fig. 1.3.6
Double frequency displacement in the case of sound reflection via a moving reflector

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

$$F_d \approx \frac{F_o \cdot V \cdot 2}{C}$$

To calculate the Doppler displacement the angle of incidence of the sound waves on the reflector is also of great importance, this is shown in the analogy in Fig. 1.3.7.

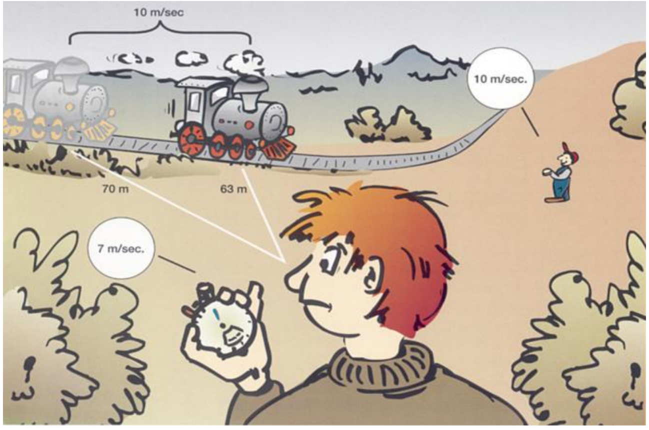

Fig. 1.3.7 – Analogy for the Doppler effect:
The size of the frequency displacement (here an approaching train) depends on the angle between the reflector (train) and the transmitter / recipient (observer)

If the train shown here moves 10 m in 1 second, the observer, who is standing at an obtuse angle to the railway tracks, merely registers the train moving towards him by 7 m. In contrast, the observer who is standing at an acute angle to the tracks registers the train moving towards him by the actual 10 m. Therefore, the Doppler displacement increases more if the angle of incidence between the sound wave and the reflector is more acute. For this reason the equation must be corrected by the cosine of the angle α between the sound ray and the reflector (longitudinal axis of the vessel). Therefore the equation now is:

$$F_d = \frac{F_o \cdot V \cdot 2 \cdot \cos\alpha}{C}$$

This fact will become increasingly important when using the Doppler probe.

If both the transmission and reflection frequency are known, the speed of the reflector can be calculated easily by rearranging the equation:

$$V = \frac{F_d \cdot C}{F_o \cdot 2 \cdot \cos\alpha}$$

We are aware of this principle from everyday life too. It is used when measuring the speed of cars via radar using electromagnetic waves.

Unidirectional Continuous Wave Doppler 2.1.

The phenomena that the frequency of the sound received is displaced relative to the sound transmitted if it has been deflected via a moving reflector, is used in Doppler technology.

As shown in Fig. 2.1.1, a transmission crystal is located in a pen shaped probe. This crystal, if excited by an oscillator, radiates a certain frequency – in our example 8 MHz. This ray of sound is reflected by the moving corpuscular blood components. According to the Doppler effect, the reflected frequency is therefore displaced up or down relative to the transmitted frequency depending on the direction of flow of the blood.

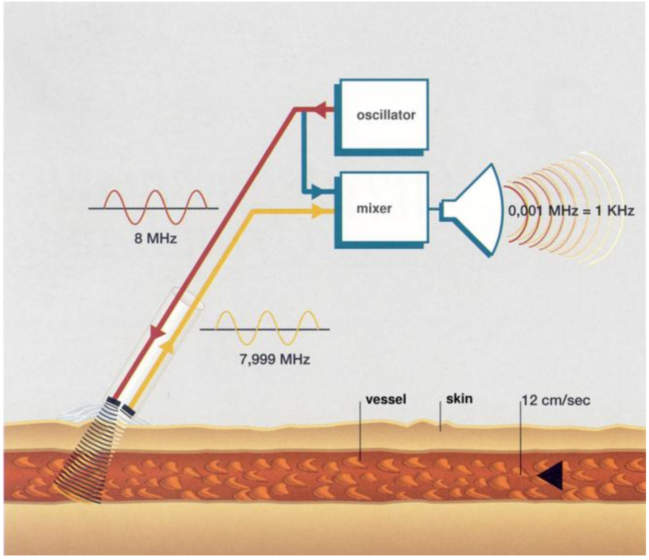

Fig. 2.1.1
Diagram to show the function of a unidirectional Doppler device

As in our example, the blood is moving away from the sonic head, the frequency is shifted down. The receiving crystal only registers a frequency of 7.999 MHz. Both frequencies are now compared by superimposing the phases on top of one another in a mixer and the amount of the Doppler displacement is formed (see Fig. 2.1.2). The difference of 0.001 MHz, i.e. 1 KHz, is in the audible range and is transmitted through a loudspeaker.

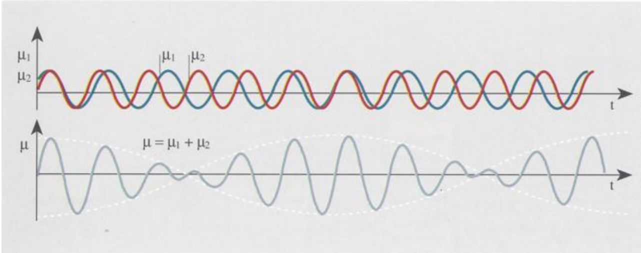

Fig. 2.1.2
Phase superimposition in the mixer and creation of a Doppler frequency by summation and subtraction of the transmission and receiving frequency

Hence, Doppler equipment provides a frequency proportional to the rate of flow of blood. The faster the flow of blood, the greater the difference between the signal transmitted and that received, the higher the sound transmitted.

If, in the example in Fig. 2.1.1, the blood were to flow with the same speed of 12 cm / second in the opposite direction, a Doppler displacement would take place which deviates upwards from the transmitted frequency – namely 8.001 MHz. If the difference is formed again, the sound created would have a frequency of 1 KHz.

Therefore, using these simple Doppler devices, it is not possible for the examining doctor to recognize whether the blood is moving away from the probe or towards the probe. For this reason these devices are known as devices which cannot recognize direction, non-directional or unidirectional devices.

Bidirectional Continuous Wave Doppler 2.2.

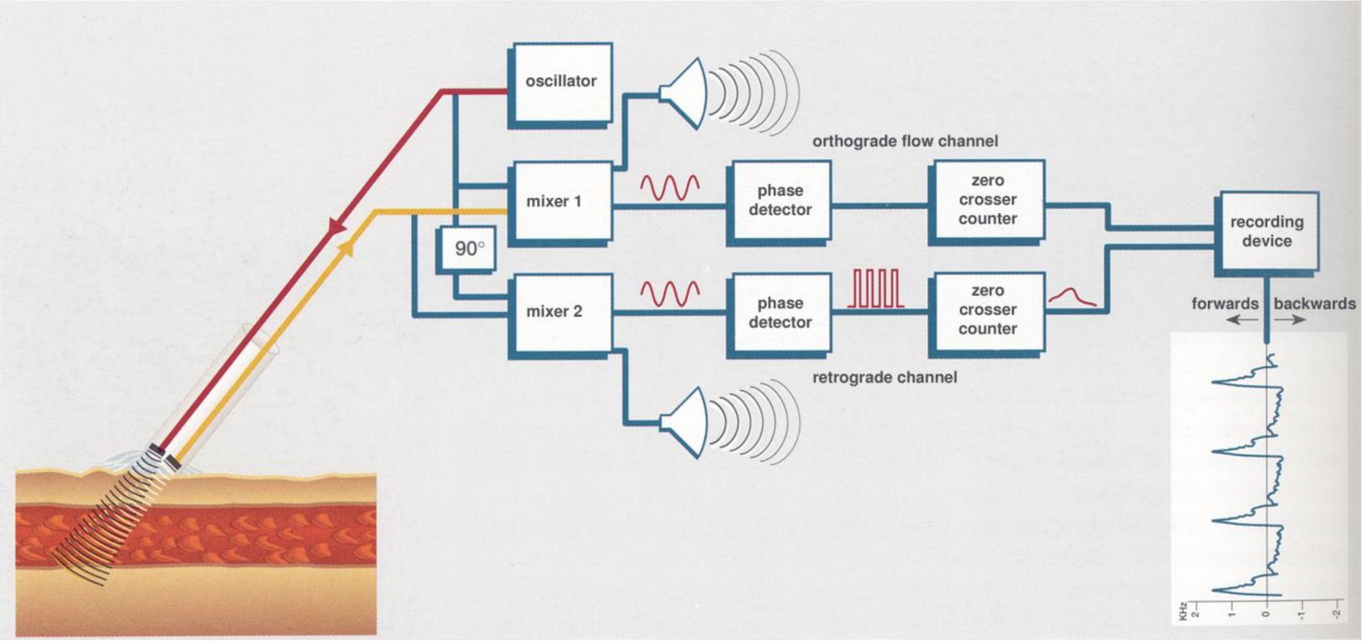

Fig. 2.2.1
Function of a bidirectional Doppler device

Due to the extremely more complicated electronics of directional or bidirectional Doppler devices compared to the unidirectional pocket devices, they are able to detect the flow of direction of blood relative to the sound probe. (Fig. 2.2.1).

Complicated devices distinguish between the direction of flow using the acoustic signal. Using stereo technology, the opposite directions of flow are transmitted through different loudspeakers.

Graphically, the direction of the flow of blood is shown as a cumulative curve on a monitor or on an integrated thermo-array printer. Orthograde flow is usually shown above and retrograde flow is shown below the zero line which is also continuously displayed. To convert the Doppler frequency into a proportional deflection, a so-called zero-crosser is used for this equipment.

The frequency of the Doppler signal is “counted“ in the simplest way possible and then converted into an analogue voltage value.

In devices which are not as technically sophisticated, correctly presenting the cumulative curve can be a problem as soon as the Doppler probe records two opposing flows which is frequently the case in the carotid flow area. Here, in addition to the signal from the common carotid artery the opposing signal from the jugular vein is recorded. In this case, the cumulative curve of the arterial signal is disturbed by the lower frequency signals of the vein and the arterial cumulative curve is displayed incorrectly (the curve is closer to the zero line than it should be).

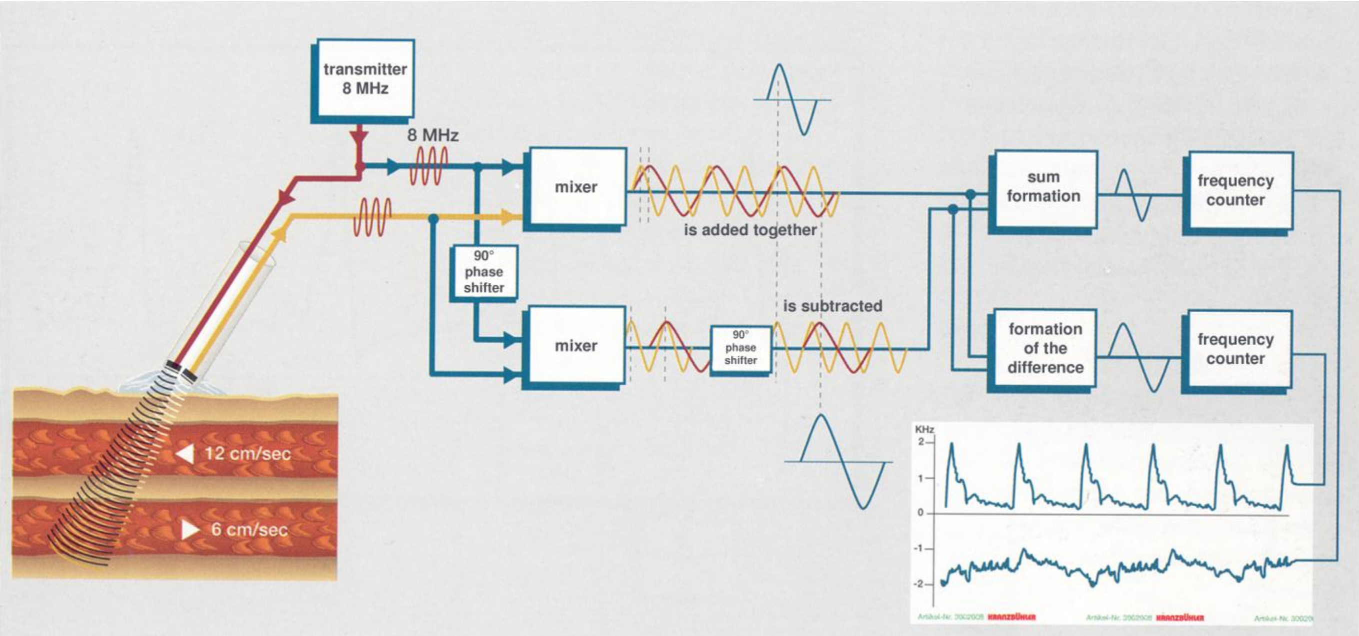

Fig. 2.2.2
Diagram showing the function of the outphaser principle

This is why the use of devices which use the so-called outphaser technology is recommended. These devices simultaneously separate forward and backward flow signals by comparing phases and pass on these signals via two separate channels. (Fig. 2.2.2).

This results in the artefact-free registration of arterial signals which have venous signals superimposed on them and vice versa (Fig. 2.2.3).

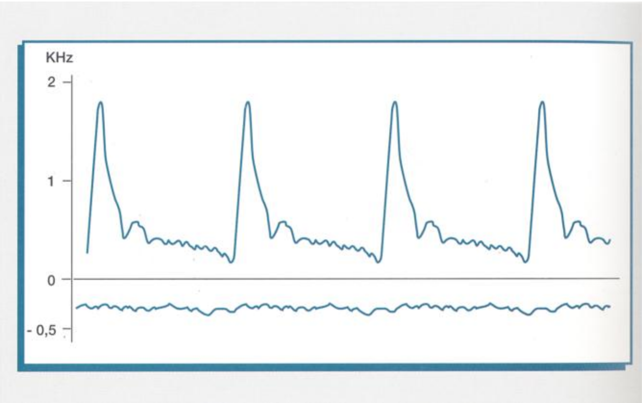

Fig. 2.2.3
Simultaneous representation of an arterial forward and venous backward flow using the example of the common carotid artery and the jugular vein

However, these devices do not take the different speeds of the erythrocytes across the cross-section of the vessel into account. Only the most frequently occurring frequency is recorded – the so-called modal frequency.

Nevertheless, these devices are well suited to qualitatively assess blood flow as long as the examiner only uses secondary parameters such as shape of the curve, direction of flow or diastolic deflection. However, they are not suitable for the quantitative assessment of a vascular process based on the height of the amplitude.

Frequency spectrum analysis 2.3.

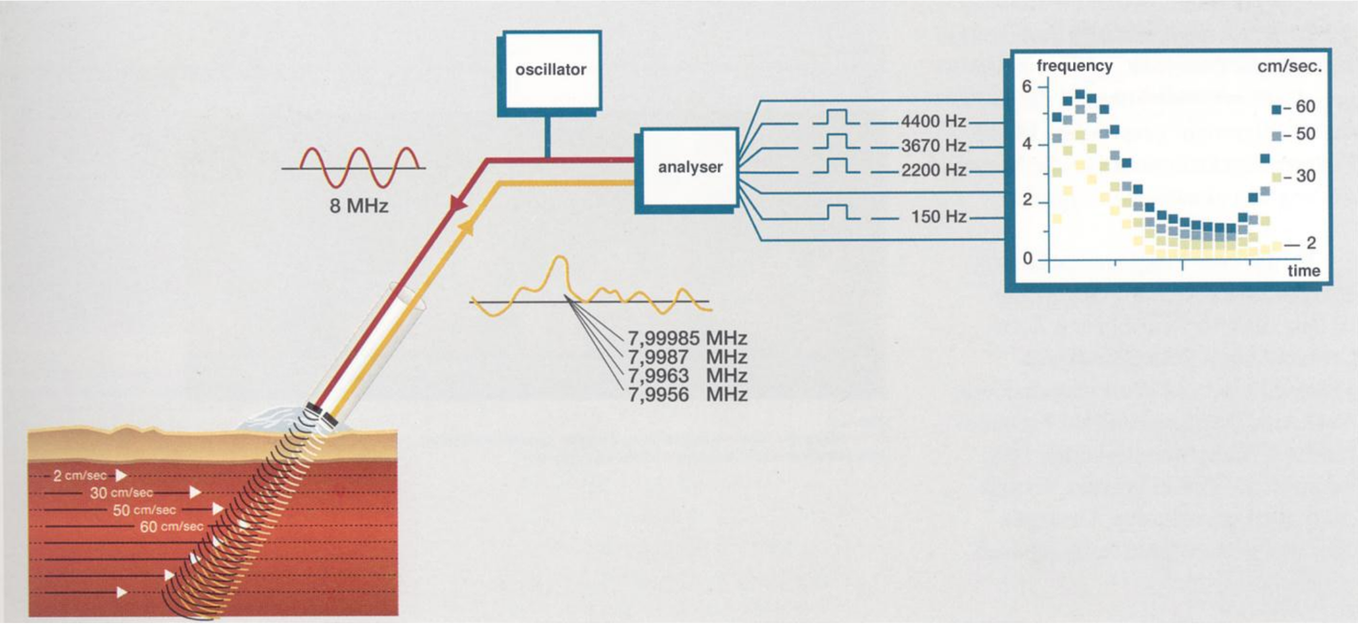

Fig. 2.3.1
Diagram showing function of frequency spectrum analysis

Frequency spectrum analysis takes account of the fact that the corpuscular blood components move through the cross-section of the vessel with different speeds. In reality, the Doppler probe does not only record one speed but a mixture of frequencies which is caused by these different speeds. This mixture of frequencies is passed on to an analyzer. This analyzer calculates the Doppler shift relative to the transmitted frequency for every frequency received. The Doppler frequencies are displayed on top of each other on a monitor as light spots on a vertical frequency and a horizontal time axis. (Fig. 2.3.1). Each point represents a specific frequency and speed.

Hence every speed causes its own Doppler curve which, superimposed on one another add up to give a spectrum (Fig. 2.3.2). In addition, a frequency spectrum analysis provides information on the frequency of occurrence of the individual speeds. This quantity of the occurring speeds is represented using the intensity or colour of the respective light spots. Speeds which occur very frequently are shown very light in the example in Fig. 2.3.3, speeds which occur less frequently are darker. In general it can be said that frequency spectrum analysis is the visualization of the acoustic signal.

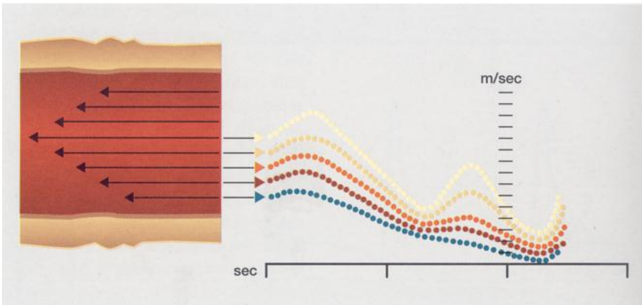

Fig. 2.3.2
Generation of a spectrum by superimposing the Doppler curves created by the individual speeds on top of each other

In contrast to the zero-crossers described above, frequency spectrum analysis can also visualize low degree stenosis very well. The frequencies and speeds shown are correct for known angles so that quantitative statements can be made regarding the degree of the stenosis.

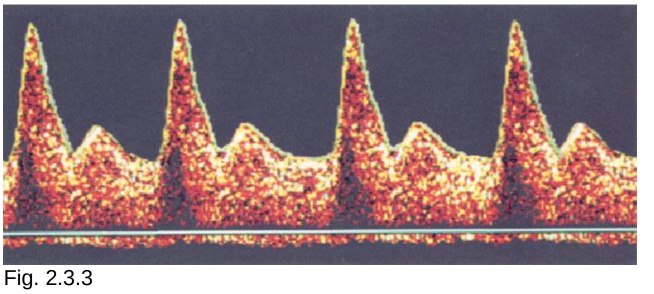

Fig. 2.3.3
Representation of the frequency of individual speeds via intensity or colour of the light spot

Pulsed Doppler 2.4.

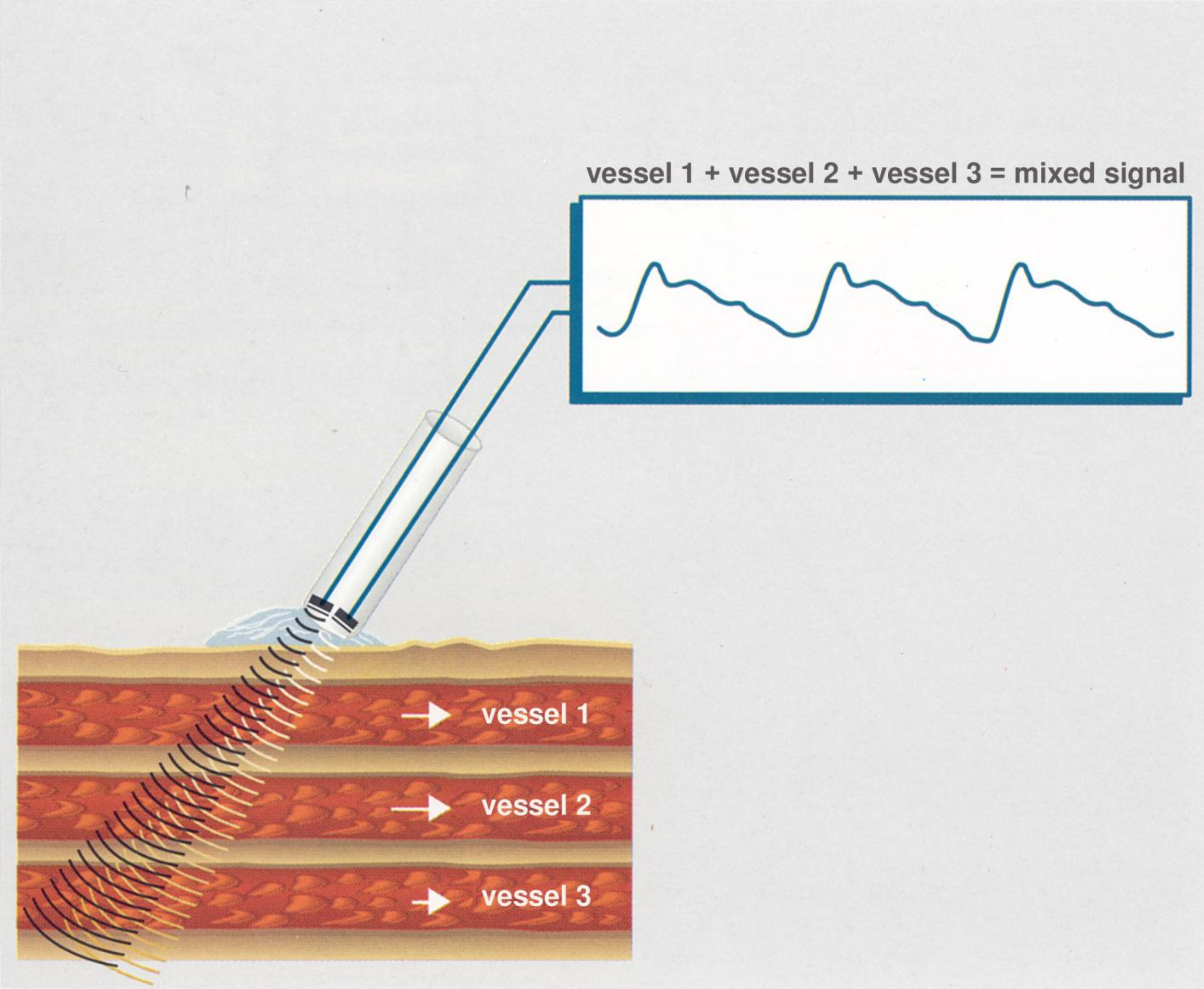

Fig. 2.4.1
In the CW process, there is no possibility to differentiate between individual vessels if a number of vessels with the same direction of flow are recorded simultaneously

The Continuous Wave Doppler technology described so far fails if the sonic beam records a number of vessels with the same direction of flow at the same time. In this case, these devices receive a mixed signal from all vessels lying along the axis of the beam. This means that you can no longer distinguish between the individual vessels (Fig. 2.4.1). In contrast, pulsed Doppler devices work with depth selection. Using this technology, it is possible to record the flow of blood within a defined, freely selected depth range.

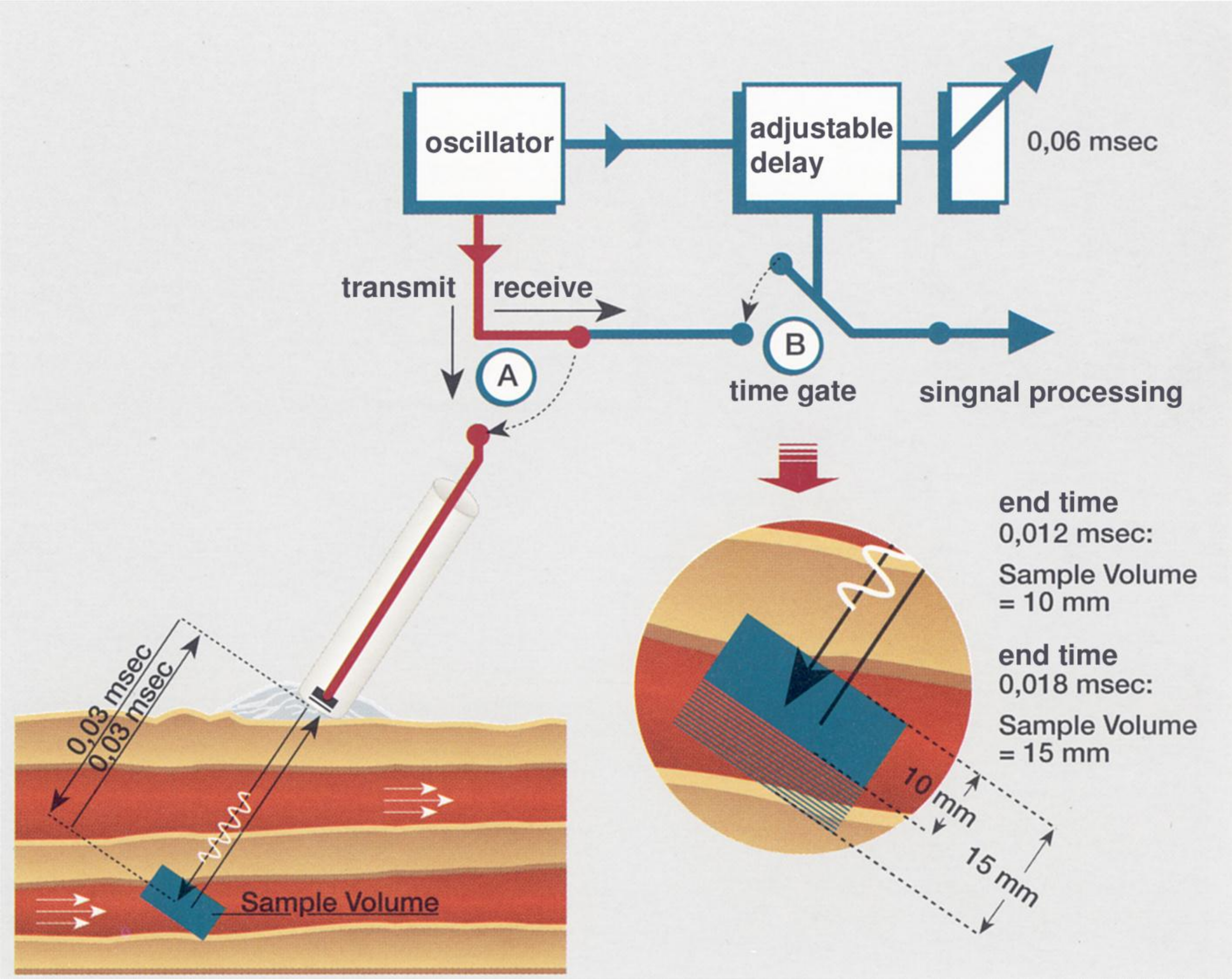

Fig. 2.4.2
Diagram showing the function of a pulsed Doppler

Fig. 2.4.2 is a diagram showing the function of a pulsed Doppler device. In contrast to the continuous wave technique, these devices work with only one crystal. This crystal serves both as a transmitter and receiver, but it does not continuously radiate ultrasonic sound. Instead it only transmits a short sound impulse, a so-called “burst” which only lasts for a fraction of a millisecond. Then using a switch (A), the crystal is switched to receive mode. Based on the known speed of sound of approx. 1.540 m/s in tissue, the device now calculates the period the impulse requires to travel to a specific depth selected by the user of the device and back to the crystal again.

At precisely this time, the time gate (B) closes for a short time. The signals from the required depth can pass. Echoes, which arrive before and after this running time, are ignored. Immediately the signals have passed the time gate, the crystal is set to transmit again and a new signal is generated.

The range from which the signals are derived is called the Sample Volume as the sound impulse expands in three dimensions. Whereas the lateral expansion of the sample volume depends on the diameter of the crystal and therefore cannot be changed, the axial expansion can be varied by varying the length of time the system is in receive mode – the length of time the time gate is closed.

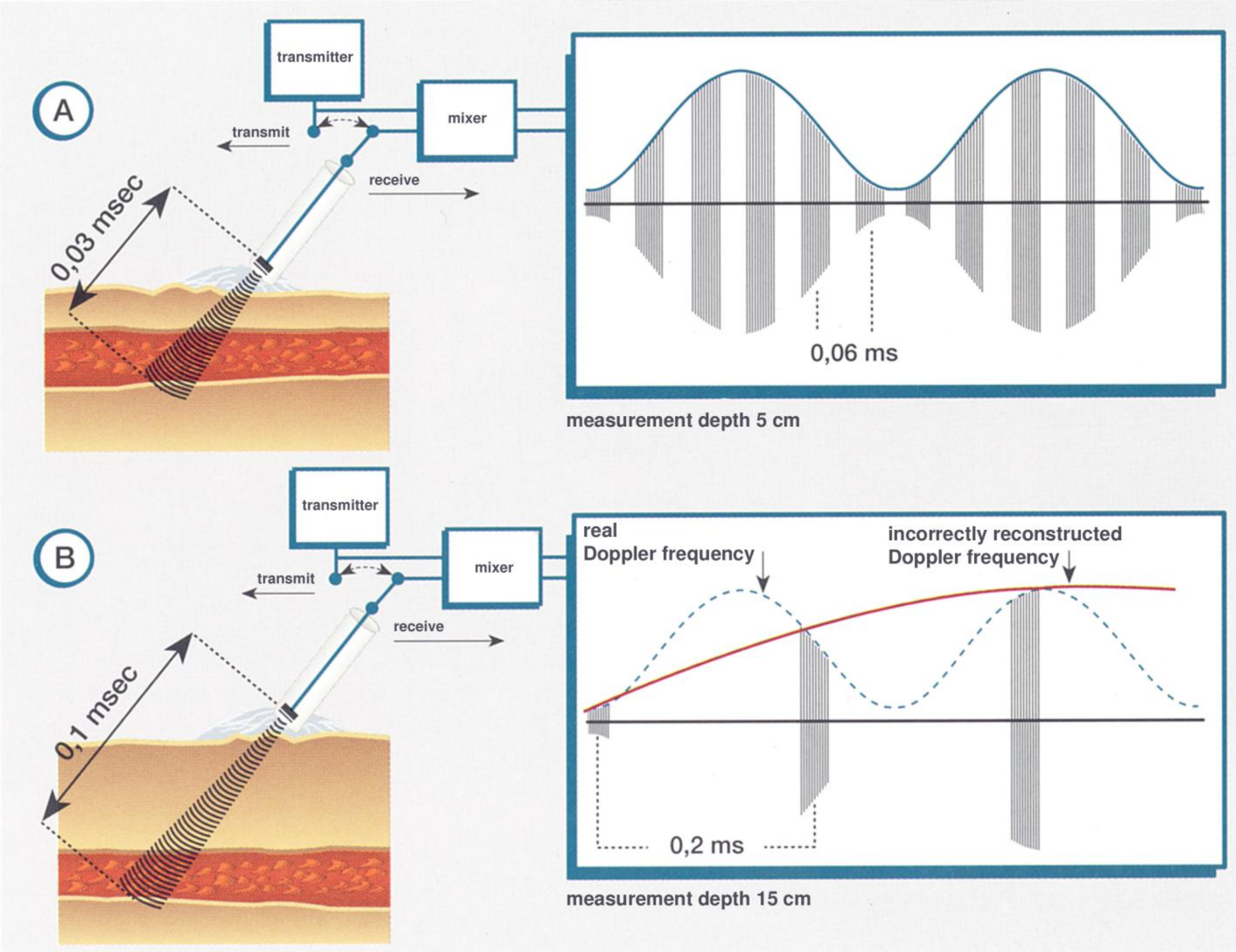

Fig. 2.4.3
Dependency of the PRF of the required measurement depth and the resulting reconstruction of the Doppler frequency

The frequency with which the impulses are transmitted is called the Pulse Repetition Frequency or PRF. This depends on the required measuring depth. The deeper the sample volume is, the greater the waiting period between the individual impulses – therefore the lower the PRF.

If, as shown in Fig. 2.4.3, measurements are made at a depth of approx. 5 cm, the time between two transmitted impulses is approx. 0.06 milliseconds. The PRF is then approx. 16.6 KHz. At a depth of 15cm, the PRF is only approx. 5 KHz, as the sound impulse requires approx. 0.2 milliseconds to travel to and from the point to be measured. Pulsed technology means that the Doppler curve is not available as a continuous signal. It is made up of individual reception segments which the device uses as a base with which to reconstruct the Doppler curve.

If the PRF is high compared to the Doppler frequency, i.e. the depth at which measurements are being made is small, it is not difficult to reconstruct the Doppler curve as a large number of received segments are available (Fig. 2.4.3 A).

In contrast, if the PRF is low compared to the Doppler frequency, an error in measuring is made generally known as Aliasing.

In this case, the wave shape has been reconstructed incorrectly as there are an insufficient number of received segments available (Fig. 2.4.3 B).

The Aliasing phenomena becomes obvious in the frequency spectrum as from a certain limit onwards, the curve is cut off and paradoxically it is displayed underneath the zero line. This limit is the maximum Doppler frequency that can be measured. It depends on the measuring depth and therefore also on the PRF and is described as the Nyquist-Limit (Fig. 2.4.4 A).

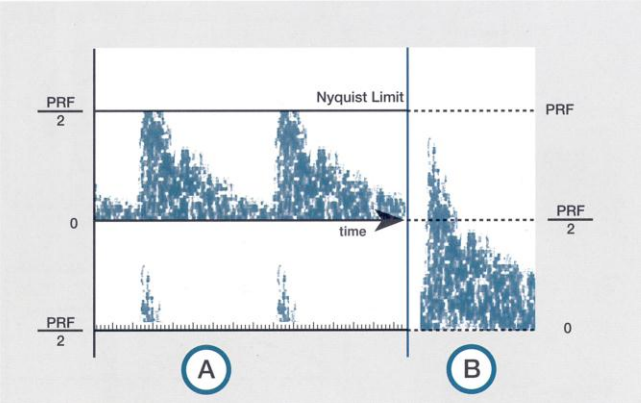

Fig. 2.4.4 (A, B)
representation of the Aliasing phenomena when the Nyquist limit (A) has been reached and correction by moving the zero line (B)

This Nyquist-Limit is always reached then when the highest measurable Doppler frequency is more than half the PRF:

$$Nyquist - Lymit = \frac{PRF}{2}$$

By moving the zero line, the representation can be corrected electronically. However, in that case only one direction of flow can be displayed (Fig. 2.4.4 B).

In addition, the Doppler frequency can be so extremely high that multiple aliasing takes place which can be observed as a wide frequency band in the frequency spectrum (Fig. 2.4.5).

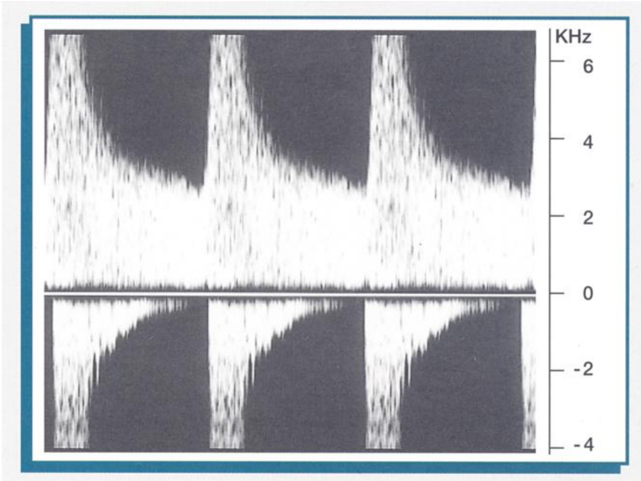

Fig.2.4.5
Diagram showing multiple aliasing during systole

As a result, pulsed Doppler devices cannot be used to measure every high speed.

The Duplex System 2.5.

The Duplex system is a combination of an ultrasound section image device and a Doppler device. In accordance with the extensive possibilities of application a large number of techniques to achieve the required results are also available. However, what they all have in common is that the sample volume can be positioned when viewed and that the Doppler spectrum can be derived at the same time alternately with the section image.

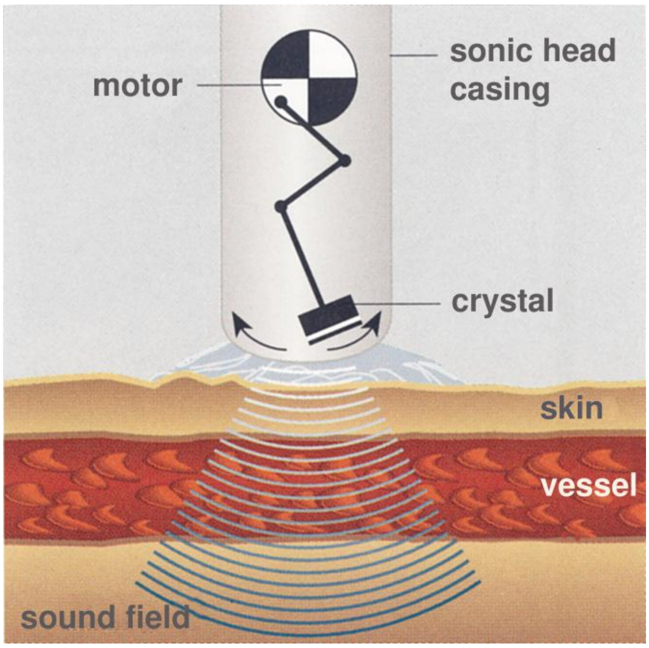

Fig. 2.5.1
Mechanical sector sound head (Wobbler)

Originally so-called mechanical sector scanners were used for this purpose. This is based on the principle that a crystal, (or a number of crystals), are moved backwards and forwards along a circular path (Wobbler, Fig. 2.5.1) or are rotated (rotation method, Fig. 2.5.2).

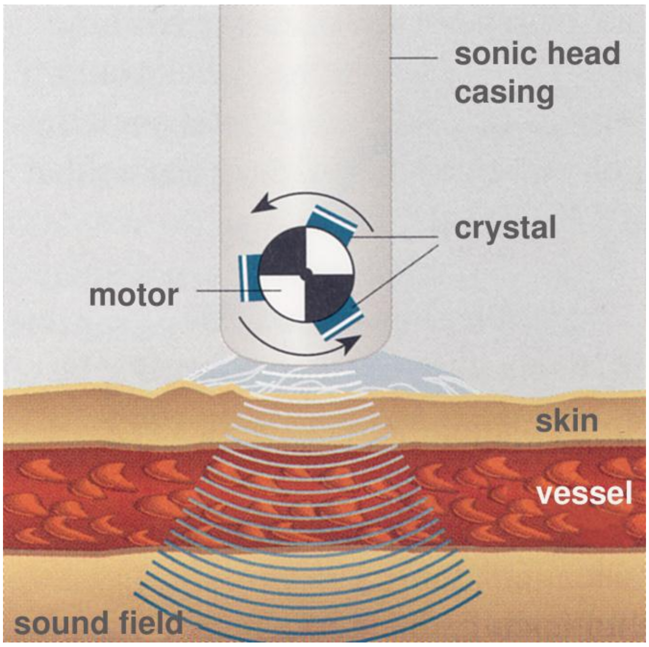

Fig. 2.5.2
Mechanical sector sound head (rotation method)

By selecting the appropriate function, a line can be shown on the ultrasound image along which the sample volume can be displaced. The sample volume is positioned in the required section of the vessel. Using a key or a foot switch, the crystal is stopped and the 2 D image is frozen. Now the crystal transmits sound impulses in pulsed mode along the axis of the inserted line and thus the Doppler spectrum is derived. (Fig. 2.5.3). In CW mode this line represents the sound beam.

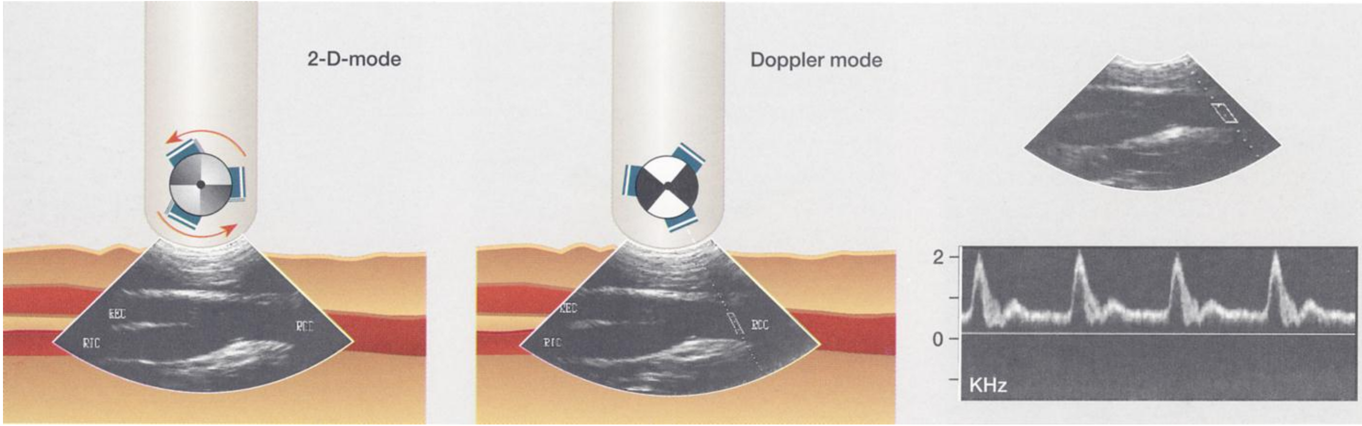

Fig. 2.5.3
Creation of a Doppler spectrum with the aid of a mechanical duplex system

The disadvantage in this case is that the 2D image and the spectrum cannot be displayed simultaneously. In addition, whilst insonating, it is possible to displace the sound head which means that the real position of the sample volume or the CW line no longer coincides with the position of the frozen 2D image displayed. In addition, these sound heads are subjected to large mechanical loads due to the constant braking of the crystal.

On the other hand, this technology provides a superb Doppler signal as the energy does not need to be used simultaneously for the 2D image and the Doppler which is the case with the wide spread array sound heads today.

Array sound heads work with a number of crystals arranged next to each other.

According to their radiation behaviour these are sub-divided into Linear Arrays (Fig. 2.5.4), Convex Arrays (Fig. 2.5.5) or Phased Arrays (Fig. 2.5.6), which are also known as electronic sector scanners. The sector shaped radiation behaviour of the phased arrays is achieved by addressing the individual crystals or crystal groups at different times. This sound head technology makes it possible to display a 2D image and a Doppler spectrum simultaneously. Now, it is possible to continuously correct the sample volume or the CW line whilst viewing. At the same time these electronic arrays are not subjected to any mechanical loads.

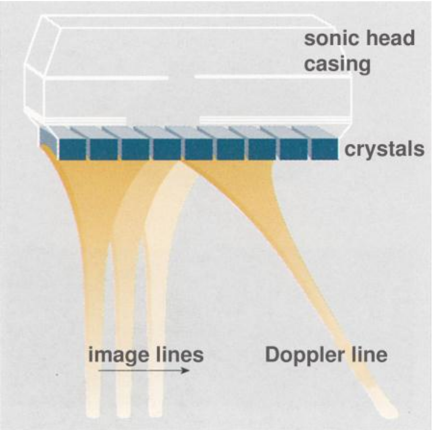

Fig. 2.5.4
Radiation behaviour of a linear array

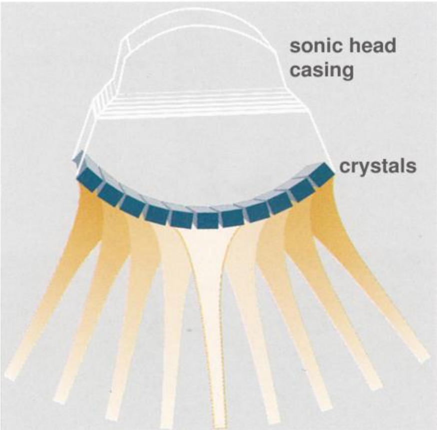

Fig. 2.5.5
Radiation behaviour of convex array

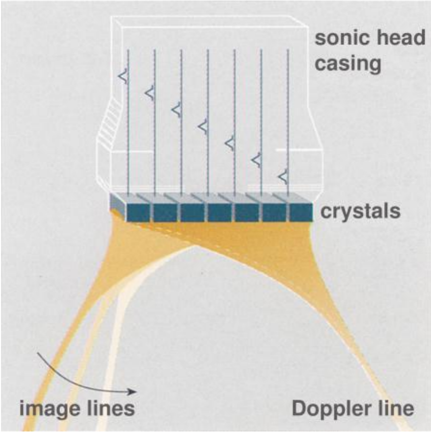

Fig. 2.5.6
Radiation behaviour of a phased array

The Colour Duplex System 2.6.

Colour coded Doppler sonography is also known as CFI (Colour Flow Imaging) or area Doppler sonography. In contrast to conventional duplex systems, colour duplex systems not only generate one sample volume but a number of sample volumes based on every ultrasound line (Fig. 2.6.1).

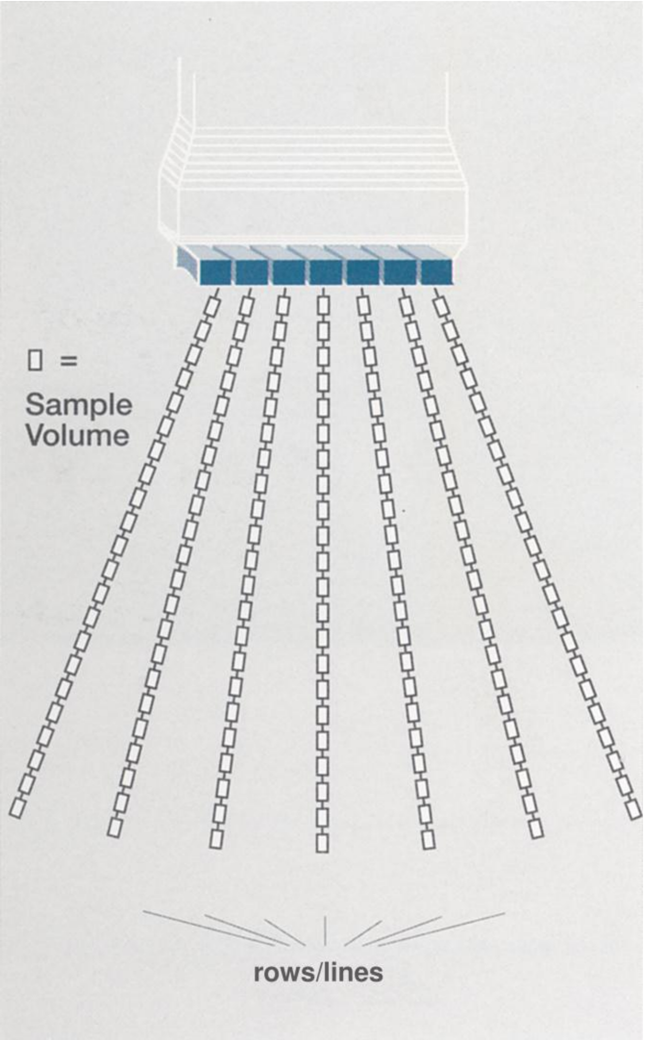

Fig. 2.6.1
Construction of a number of sample volumes of a CFI phased array sonic head

In addition, the Doppler signals received at the same time with the 2D information are not displayed as a frequency spectrum but as colour coded information in the correct position in the section image. Blood flow which is coming towards the sonic head is usually coded red and blood flow moving away from the sonic head is usually coded blue (Abb. 2.6.2).

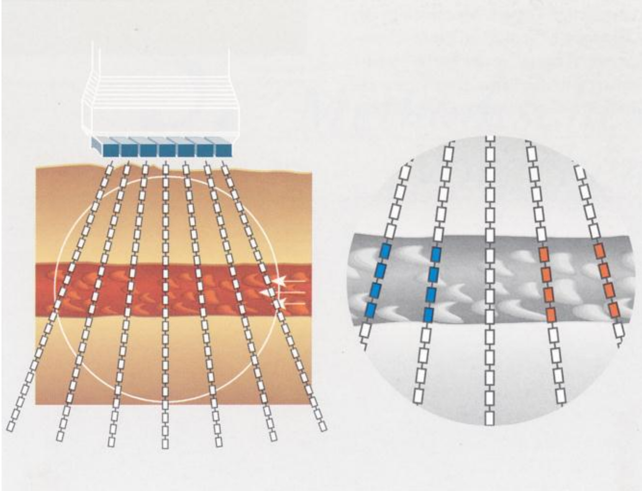

Fig. 2.6.2
Colour coding in a colour duplex system dependent on the direction of flow

This means that the examiner is given a rapid overview of the haemodynamic conditions in the area being investigated. However, the colour permits no quantitative statements to be made.

For this reason the additional representation of the frequency spectrum is still absolutely essential.

To do this, in the same way as conventional duplex systems, the sample volume in the so-called “region of interest“ can be positioned in colour and the Doppler spectrum can be derived also (Fig. 2.6.3).

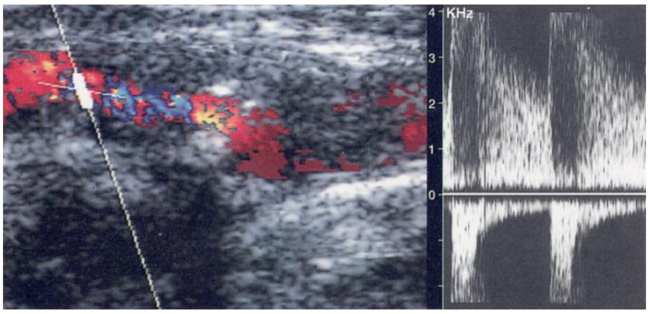

Fig. 2.6.3
Creation of a frequency spectrum from the colour image (high degree stenosis of the internal carotid artery)

The correct transmission frequency 3.1.

In principle, the ultrasound energy transmitted is absorbed by the tissue and partially reflected. Therefore the intensity of the sound decreases continuously with increasing depth. The shorter the wavelength, the higher the transmission frequency and the greater the loss in intensity. Therefore it follows that as transmission frequency increases, the penetration depth of the sound decreases.

Hence the working range of a

8-10 MHz probe is approx.	8-25 mm
4-5 MHz probe is approx.	15-50mm
2 MHz probe is approx.	30-120mm

To be able to optimally subject the vessels at different depths to ultrasound, the use of different transmission frequencies is recommended. For vessels near the surface such as the posterior tibial artery, the dorsal pedal artery, the radial artery, the orbital arteries as well as veins close to the surface, we recommend the use of an 8 MHz probe.

Arteries which are further away from the surface such as the femoral artery, popliteal artery, subclavian artery, the carotid flow area as well as for investigations of deep venous systems, one usually uses a 4 MHz probe. To perform ultrasound investigations of intracranial vessels only a 2 MHz probe in pulsed mode can be used.

2 MHz probes are also used in connection with Foetal-Pulse-Detectors in order to prove foetal heart activity in early pregnancy. However, these probes work in Continuous-Wave mode.

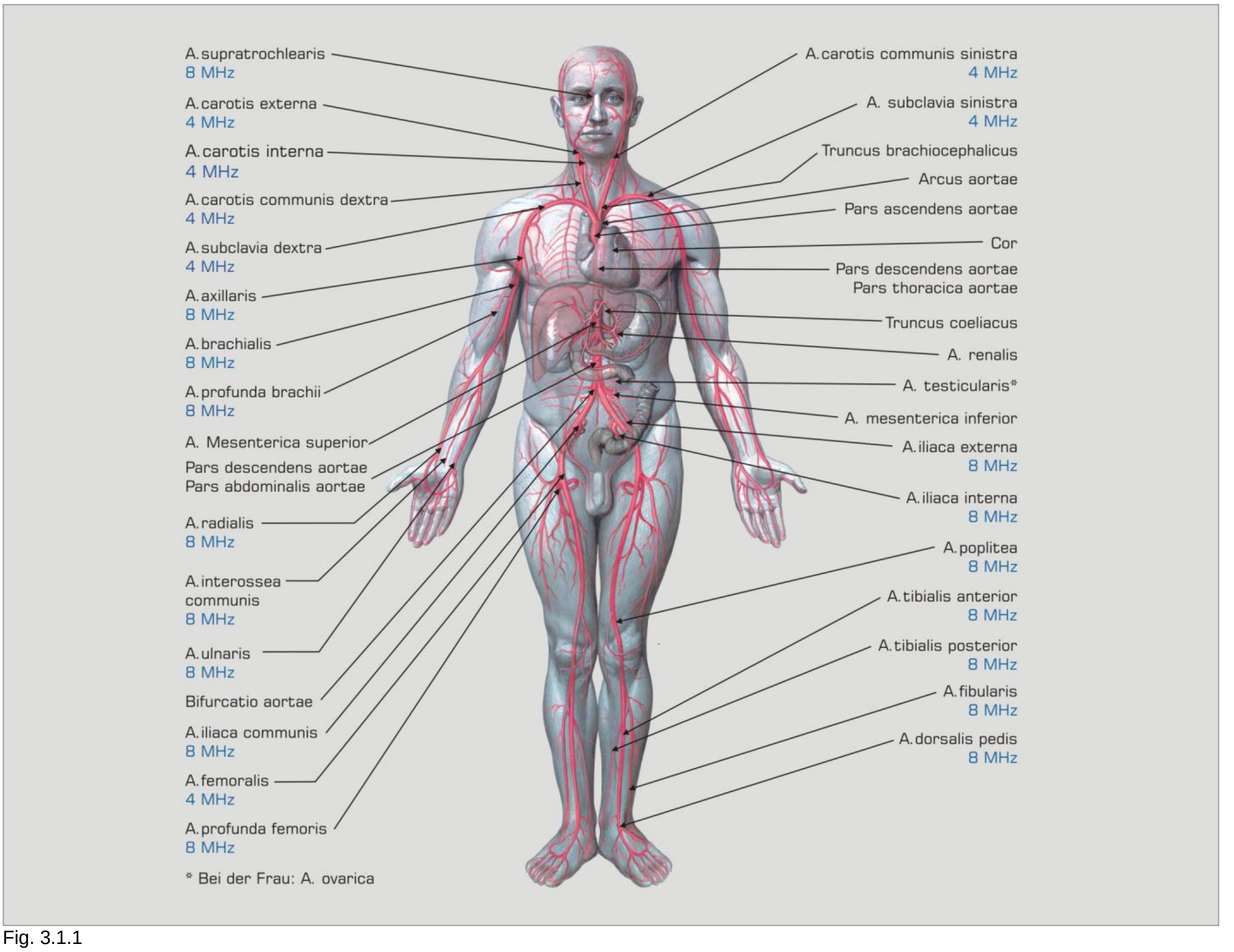

Fig. 3.1.1

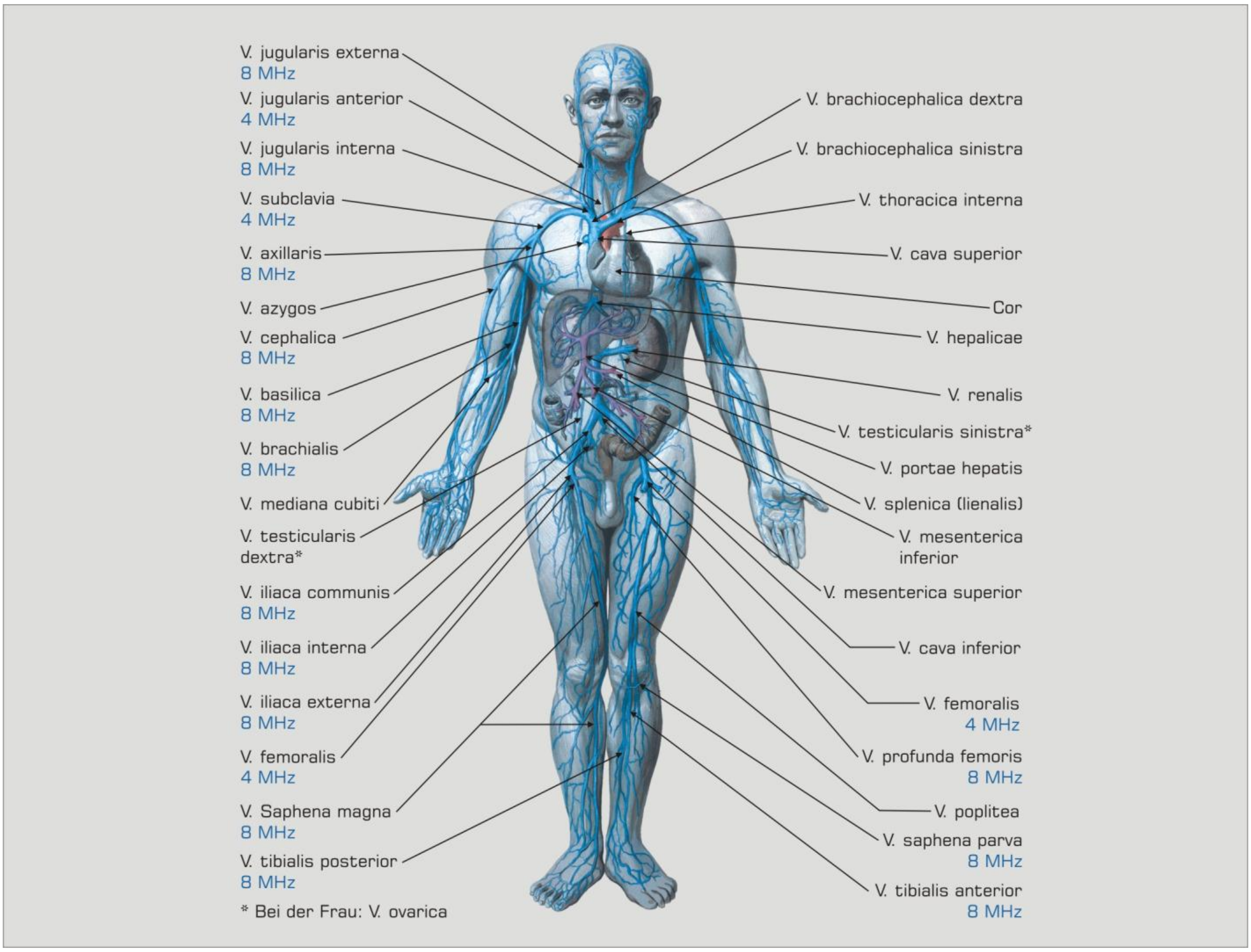
V. jugularis externa
8 MHz
V. jugularis anterior
4 MHz
V. jugularis interna
8 MHz
V. subclavia
4 MHz
V. axillaris
8 MHz
V. azygos
V. cephalica
8 MHz
V. basilica
8 MHz
V. brachialis
8 MHz
V. mediana cubiti
V. testicularis dextra*
V. iliaca communis
8 MHz
V. iliaca interna
8 MHz
V. iliaca externa
8 MHz
V. femoralis
4 MHz
V. Saphena magna
8 MHz
V. tibialis posterior
8 MHz
* Bei der Frau: V. ovarica
V. brachiocephalica dextra
V. brachiocephalica sinistra
V. thoracica interna
V. cava superior
Cor
V. hepalicae
V. renalis
V. testicularis sinistra*
V. portae hepatis
V. splenica (lienalis)
V. mesenterica inferior
V. mesenterica superior
V. cava inferior
V. femoralis
4 MHz
V. profunda femoris
8 MHz
V. poplitea
V. saphena parva
8 MHz
V. tibialis anterior
8 MHz

Fig. 3.1.2

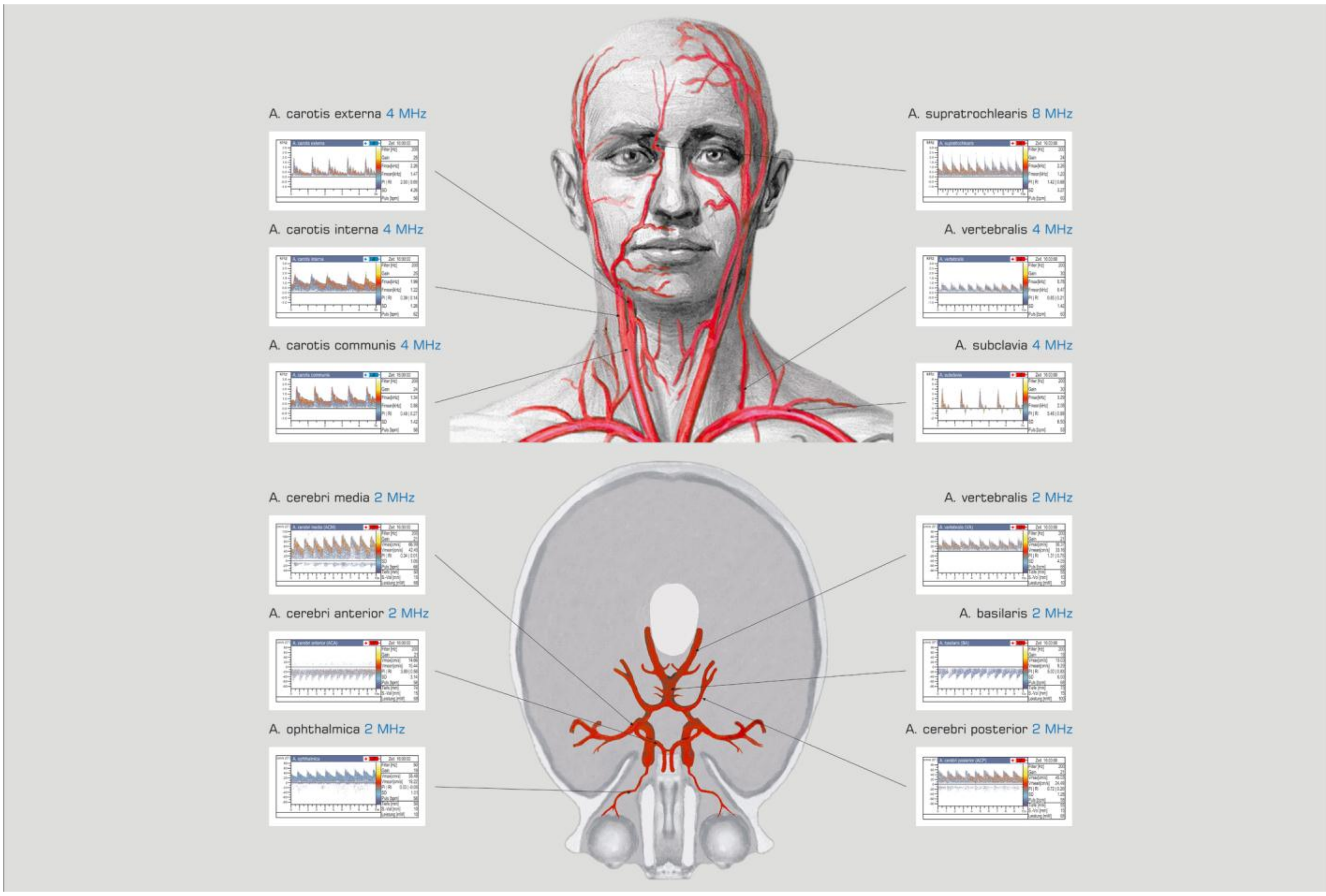
A. carotis externa 4 MHz
A. carotis interna 4 MHz
A. carotis communis 4 MHz
A. supratrochlearis 8 MHz
A. vertebralis 4 MHz
A. subclavia 4 MHz
A. cerebri media 2 MHz
A. cerebri anterior 2 MHz
A. ophthalmica 2 MHz
A. vertebralis 2 MHz
A. basilaris 2 MHz
A. cerebri posterior 2 MHz

Fig. 3.1.2

Positioning the probe 3.2.

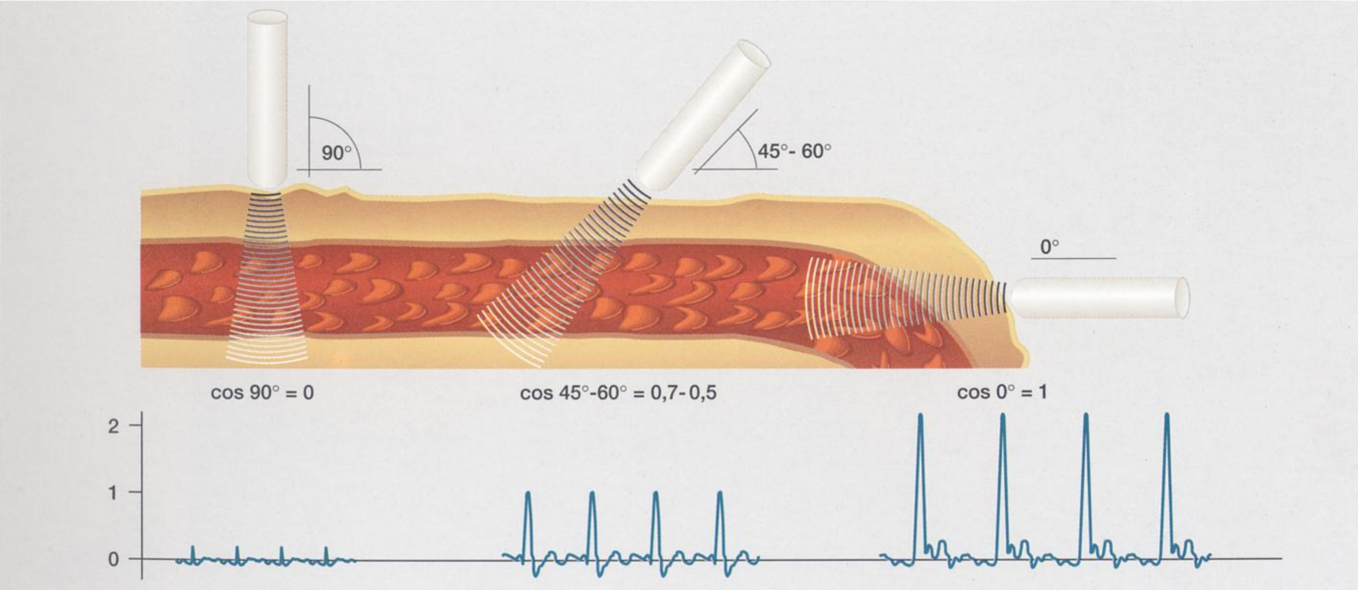

Fig. 3.2.1
Changing the Doppler shift dependent on the angle of incidence

After the correct transmission frequency has been selected, ultrasound contact gel is put on the probe which is then positioned above the vessel to be examined. The contact gel acts like a coupling medium as there must be no air between the surface of the skin and the probe as air acts as a total reflector. Under no circumstances should an electrode cream be used as a coupling medium as this can irreparably damage the probe.

When the probe is positioned you should make sure that there is an angle of between 45° and 60° between the probe and the surface of the skin. As the vessels frequently run parallel to the surface of the skin, you can expect a fairly good signal immediately. As already described, the angle between the longitudinal axis of the vessel and the ultrasound beam is very important as the frequency displacement increases the more acute the angle is between the two axes. If the probe is positioned at a 90° angle above the axis of the vessel, the frequency displacement will be virtually non-existent as the cosine of the angle 90° = 0 (Fig. 3.2.1). If the vessel is represented as a noise sample above the Doppler, the position of the probe is optimized by maintaining the position on the skin but rotating the probe in its axial and tangential angle to the axis of the vessel. (Fig. 3.2.2). The position is regarded as being optimal when the acoustically most pure Doppler sound is seen.

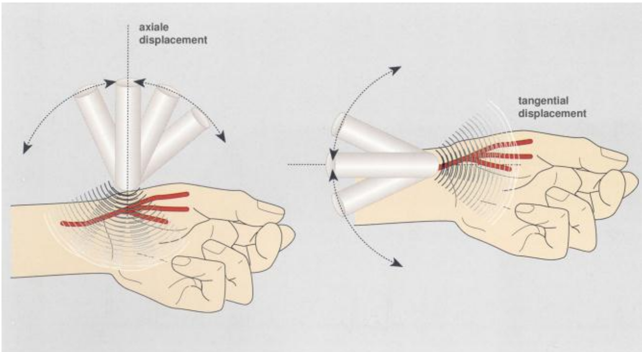

Fig. 3.2.2
Finding the optimal probe position by displacing it axially and tangentially

As Doppler sonography is an acoustic means of diagnosis, the graphical representation only serves for documentation purposes and as a visual aid. The optimum probe position should be therefore primarily based on the acoustic signal.

The sound characteristics 3.3.

As veins have a relatively steady blood flow, this noise is heard as a virtually constant howling sound.

In contrast, arteries are subjected to strong pulsations. Therefore, during the examination, during systole quite a high tone is heard due to the increased blood flow, the pitch of this note decreases during diastole.

In the case of an undisturbed laminar flow, the Doppler signal sounds clear and pure. In the area of vessel sections which show a stenosis, blood flow is characterized by strong acceleration and turbulences. Hence, the signal in the region of a stenosis sounds rough and grating (walking on gravel).

Diagnosis using Unidirectional Dopplers4.1.

Doppler blood pressure measurement

The most important application of unidirectional devices when examining peripheral arteries is Doppler blood pressure measurement to assess peripheral, arterial perfusion.

Doppler blood pressure measurement is regarded as one of the simplest ways to determine whether there are any obstacles preventing the flow of blood in the arteries of the extremities. As every haemodynamic stenosis - with the exception of simultaneous medisclerosis – results in a decrease in pressure in the respective artery after the stenosis, blood pressure should be measured as distally as possible.

Fig. 4.1.1
Insonation of the A. dorsalis pedis to measure the blood pressure via the anterior tibial artery

Usually, the examiner positions a blood pressure cuff above the ankle joint. Using ultrasound contact gel, the dorsal pedal artery in the area of the back of the foot (Fig. 4.1.1) in the extension between the 1st and 2nd toe and then the posterior tibial artery dorsal of the inner ankle is searched for (Fig. 4.1.2). As described in section 3, the signal is optimized. Then the cuff is quickly pumped up to values above the systolic values until the pulsatile noise stops.

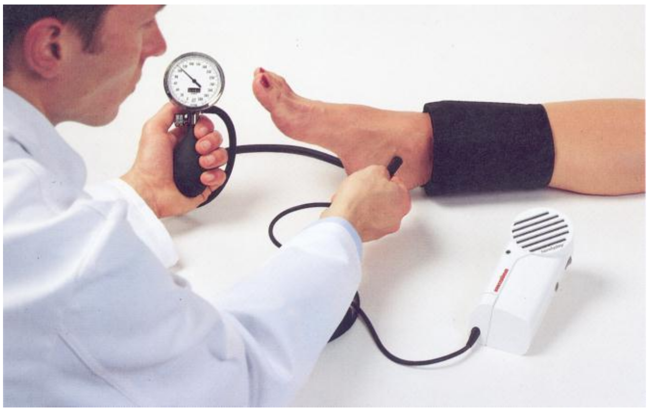

Fig. 4.1.2
Examination of the posterior tibial artery

Now the pressure is released slowly. When the first audible sound is heard, the systolic value is read on the manometer. This value is equivalent to the pressure where the cuff is. As a reference value, the pressure is measured above the brachial artery on the upper arm (Fig. 4.1.3).

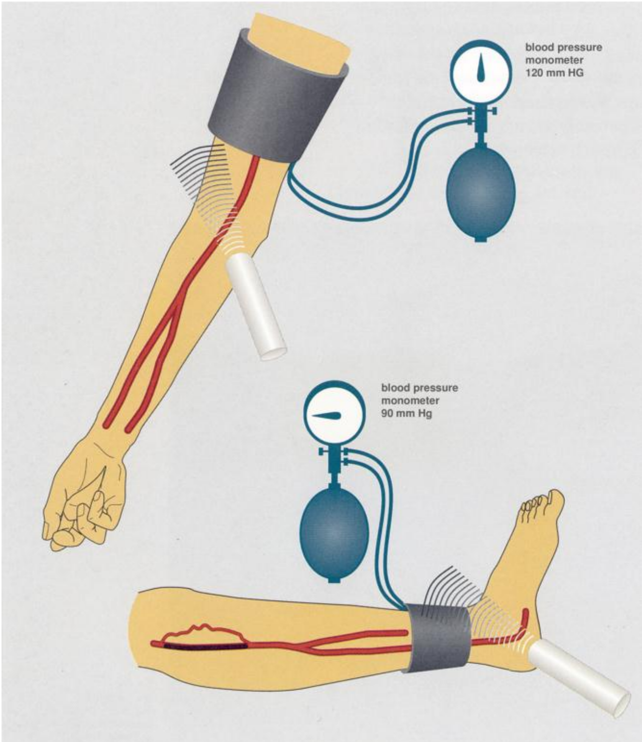

Fig. 4.1.3
Doppler blood pressure measurement (modified according to Rudofsky)

We recommend that measurements are made on both sides, on the one hand to take into account the increased value and on the other hand to obtain indications for a possible subclavian stenosis.

To assess the degree of severity, we recommend that a Doppler index is created. To do this, you must calculate the quotient using the ankle pressure and the upper arm pressure:

$$\frac{\text{ankle pressure}}{\text{upper arm pressure}} = X$$

Normal values lie between 0.9 and 1.2. Values below 0.9 are always pathological. Values below 0.5 indicate a critical circulation problem.

This way it is possible to quickly prove stenotic processes but without actually having located them. To locate them, we recommend separate measurements of blood pressure on the thigh and calf and the upper and lower arm, in which case additional multi-level processes may be discovered.

In the case of patients with high blood pressure, we recommend that a pressure gradient is also formed as despite post stenotic reductions in blood flow, high ankle pressure values are recorded here. To do this, calculate the difference in pressure between the brachial artery and the posterior tibial artery or the dorsal pedal artery. Healthy people always have a negative gradient. A difference of 10 mmHg indicates peripheral occlusive arterial vascular disease.

In the same way, using Doppler pressure measurements, approximately 50% of stenoses can be found without the typical symptoms of pathological changes having taken place already.

In this case, the ankle pressure must be measured after exercise (standing on tiptoes). Slight stenoses cause a peripheral decrease in pressure. The relative decrease in pressure is compared with the initial value as well as the period of time required for the values to rise to the initial values, the so-called return time. A decrease in pressure is regarded as being pathological. In healthy persons, the return time is less than 2 minutes.

Diagnosis using Bidirectional Doppler 4.2.

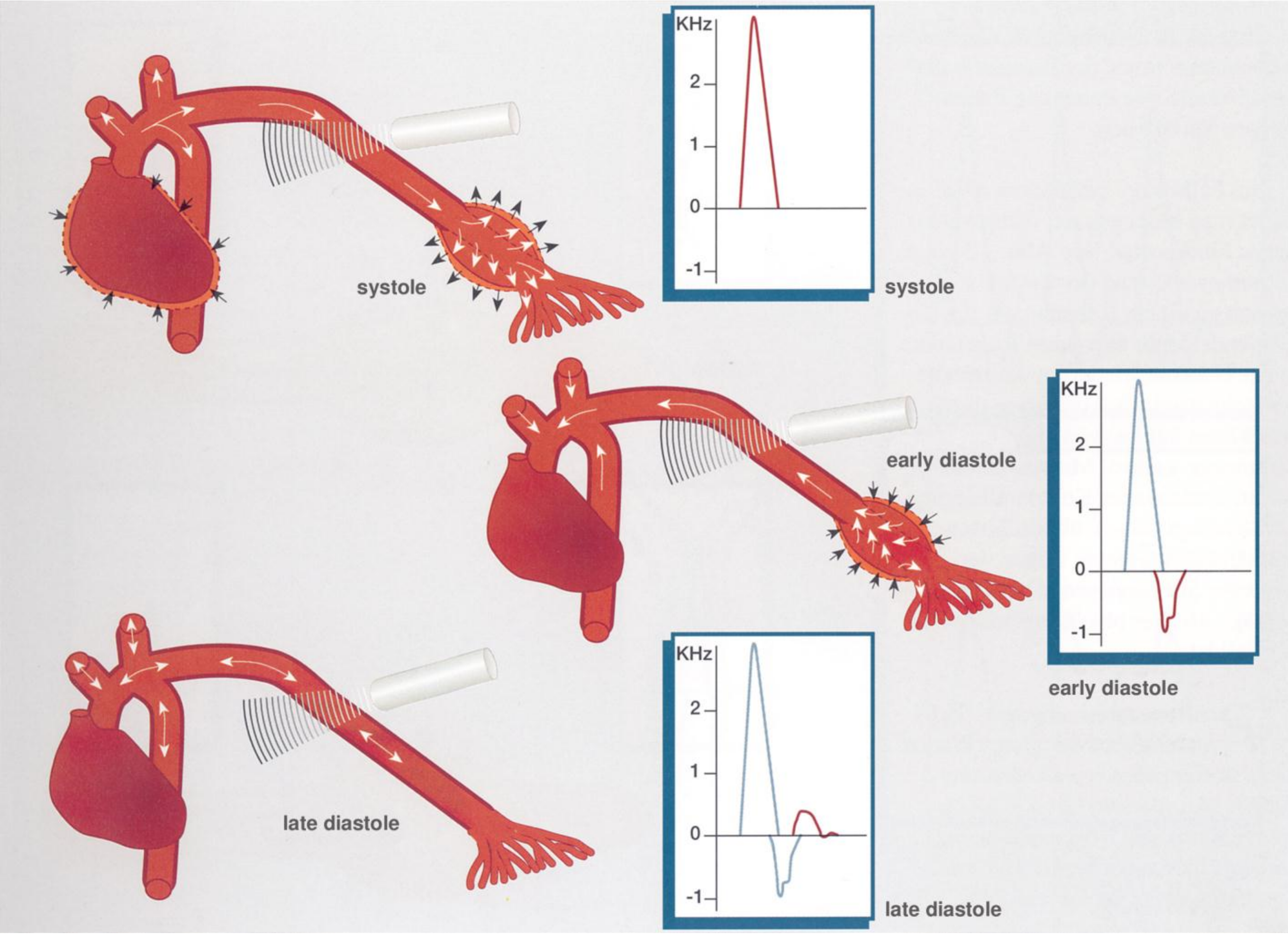

Fig. 4.2.1
Physiological flow behaviour in the arteries of the extremities

Bidirectional Doppler are more suitable for more detailed diagnostics especially to locate the position of occlusions. In addition to the acoustic signal, the curve form recorded by these devices is also analysed.

Fig. 4.2.1 shows the physiological flow conditions in the large arteries of the extremities. During systole, a large amount of blood is pressed into the peripherals through the aorta. This systolic phase is registered as an upward deflection by the Doppler device. As the blood cannot flow into the peripherals unhindered due to the decreasing vascular tonus, but on the other hand is opposed by a high ventricle pressure, blood stagnates in the distally located arteries which causes the walls of the vessels to stretch. When the ventricle pressure decreases a short time later during the early diastolic phase, the walls of the vessels which had been under pressure, return to their original position. This causes a small amount of blood to flow back in the direction of the heart.

On the device, this early diastolic retrograde flow is shown as a short deflection downwards. This deflection describes the early diastolic backward flow. Due to the elasticity of the aorta, the blood is now accelerated towards the peripheral vessels and this is registered as a second orthograde deflection. During late diastole the flow flattens out towards zero due to the intra-luminar friction resistance as well as the peripheral resistance and this results in a late diastolic flow standstill.

The flow pattern described here with an obvious early diastolic backward flow is found physiologically in the

subclavian artery,
brachial artery,
femoral artery,
popliteal artery,

as well as ankle arteries at normal temperature and a previous suitable rest phase.

However, this flow pattern depends on the length of the aorta as well as the elasticity of the vessel walls and therefore varies from patient to patient.

In the case of a peripheral AVK, the flow pattern is completely different. As shown in Fig. 4.2.2, during systole, pre-stenotic blood stagnation occurs which stretches the walls of the vessels, when the vessels return to their original size during the diastolic phase; this triggers the early diastolic backward flow due to a small amount of blood being accelerated towards the heart. With the exception of an already existing collateral vascular bed, this early diastolic backward flow component is more strongly defined due to the greater amount of stagnating blood than is usually physiolocally the case.

Due to the lack of oxygen, the post-stenotic vessels are widely dilated. For this reason blood does not stagnate and the vessels do not return to their original size.

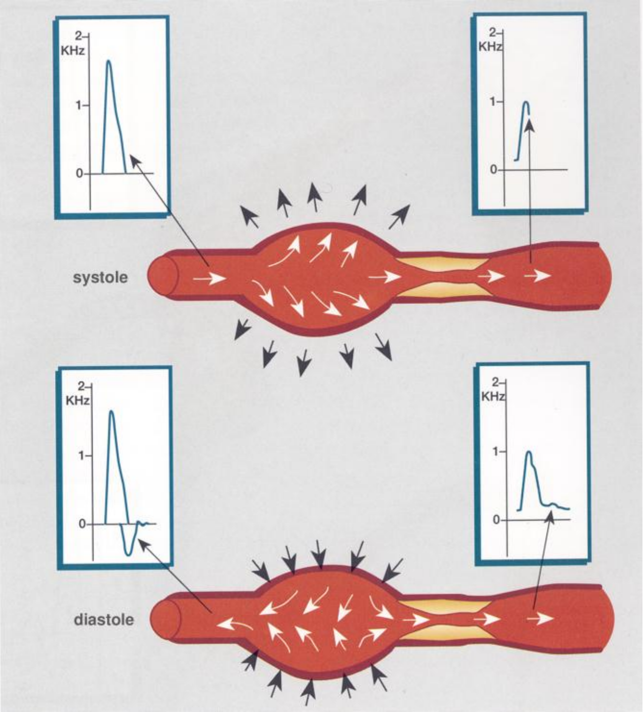

Fig. 4.2.2
Representation of disturbed haemodynamics due to a stenosis and the resulting curve

Therefore, in this case, no early diastolic backward flow component can be formed. Instead, during the diastole phase, a large volume of blood is pressed from the pre-stenotic stagnation area through the bottleneck which results in a significant post-stenotic increased diastolic flow. Due to the sudden increase in the lumen after the stenosis, the flow of blood slows down during the systolic phase in the post-stenotic region. Therefore the height of the amplitude of the systolic peak is considerably lower than that of the pre-stenotic or physiological curve. (Fig. 4.2.3).

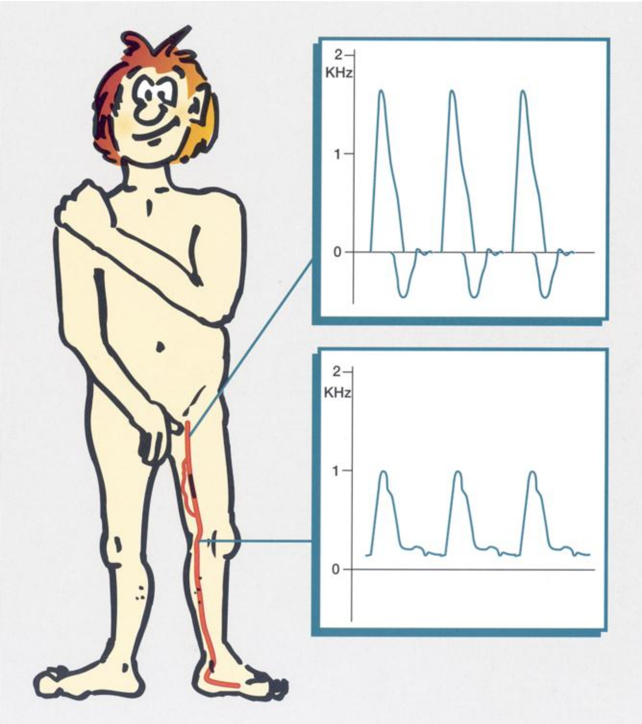

Fig. 4.2.3
Typical changes to the flow curve in an artery of the extremities due to a haemodynamically effective stenosis

In addition, in the case of an already existing collateral vascular bed (collateralisation) there is an obvious expansion of the systolic deflection, the extent of which correlates to the length of the respective collateral vessel.

If a stenotic area is subjected directly to ultrasound, the amplitude height increases significantly as the erythrocytes are accelerated strongly here. Turbulences occur post-stenotically due to the detachment syndrome, these are shown below the zero line.

Diagnosis using Bidirectional Doppler 5.1.

Subjecting extracranial cerebral blood vessels to ultrasound comes under the domain of Doppler sonography. In the industrial states, lesions of cerebral blood vessels come third in the list of causes of death. As a result, the discovery of extracranial vascular disease is incredibly important in terms of socio-medical aspects especially considering the resulting high rate of invalidity. Today, the Doppler examination of the cervical vessels is standard.

A distinction is made between the direct Doppler examination of the carotid flow area and direct insonation of the curves in the neck area, and indirect examination via the collateral circulation of the orbital arteries.

Direct Application of Sound to the Carotid Flow Area 5.1.1.

Direct Doppler examination of the carotids is usually quite simple if the physiological curves and acoustic signals are known. In addition, the characteristic noise and curves of the individual carotids differ significantly from one another.

On principle, in contrast to the peripheral arteries, extracranial cerebral vessels are not subjected to any muscular resistance worth mentioning. For this reason all vessels are characterized by a relatively high diastolic flow. (Fig. 5.1.1.1).

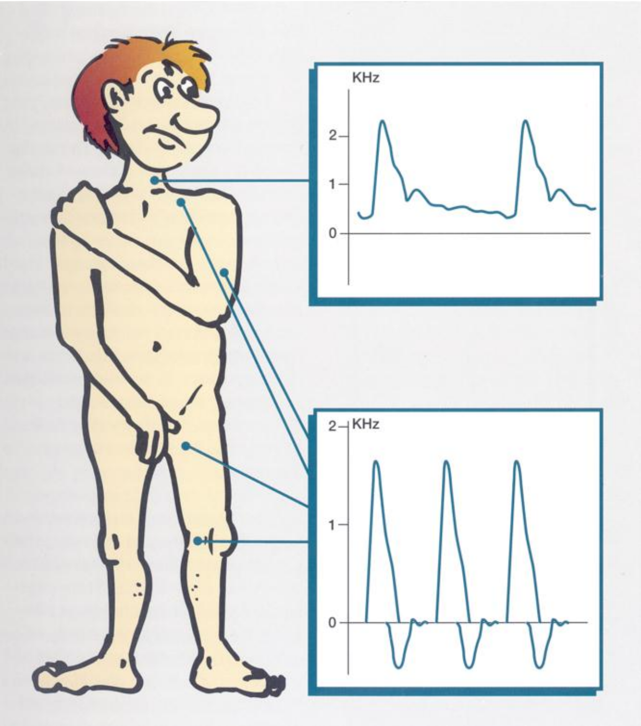

Fig. 5.1.1.1
Representation of different flow patterns of extracranial arteries supplying the brain and peripheral arteries

Fig. 5.1.1.2 shows the anatomical relationship in connection with the typical shapes of the curves.

The signal from the external carotid artery comes closest to the peripheral flow pattern. As it supplies the face, head and neck muscles, it faces the greatest resistance of all three. It is characterized by a high systolic deflection which sounds very much like a whiplash and pulsates strongly. During the diastolic phase, it decreases to relatively low flow speed values which are usually above the zero line. In some cases, the external carotid artery also shows a slight early diastolic backwards flow.

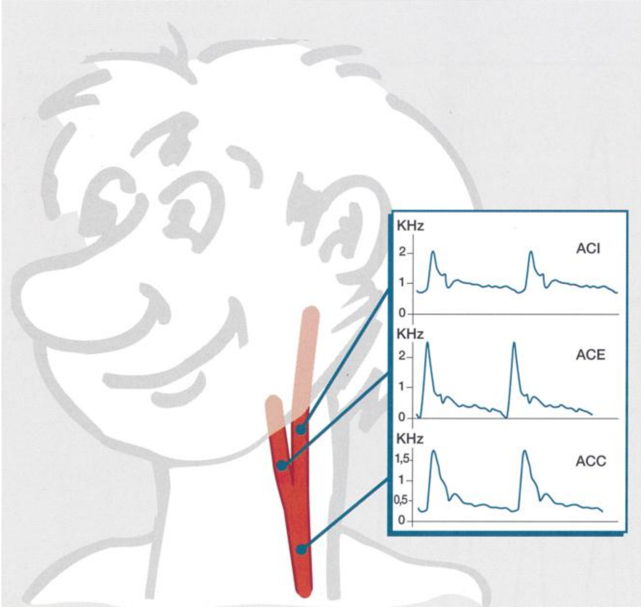

Fig. 5.1.1.2
Anatomy of the carotids and their physiological flow pattern

In contrast, the signal produced by the internal carotid artery has a very high diastolic flow due to the very low peripheral resistance. Systolically, it has similar deflections to that of the external carotid artery. As the increase starts from a very high diastolic level, it sounds very soft and rounded.

The signal of the common carotid artery is characterized by its lower systolic deflection. The diastolic level lies between that of the internal carotid artery and that of the external carotid artery.

It is easy to distinguish between the carotids both acoustically and visually. In practice the proximal section of the common carotid artery is examined with the Doppler device first, the examiner then travels along this vessel continuously. The bifurcation is recorded as a noise with a lower frequency due to the bulbous type dilatation of the vessel. After moving a little further in the direction of the head, you can hear the whiplash sound of the external carotid artery. By moving the probe in a dorso-lateral direction, in 80 to 90% of all cases, the examiner will be able to find the internal carotid artery.

Haemodynamically effective stenoses are represented quite impressively. According to Bernoulli's law, a liquid that flows through a pipeline system with different cross-sections is speeded up in the area where the lumen size decreases. At the same time the pressure is reduced in this area as the same volume must always flow during the same amount of time (Fig. 5.1.1.3). At every position in the pipeline system, the sum of speed (squared) and interior pressure is identical.

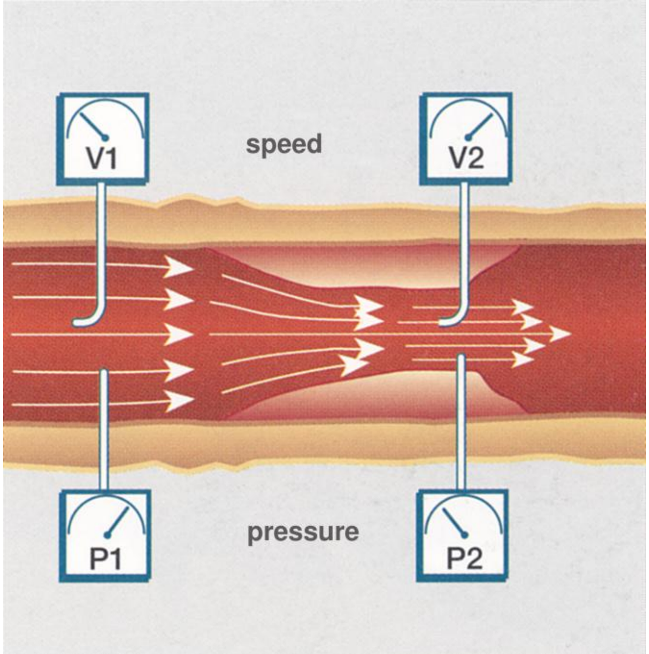

Fig. 5.1.1.3
Mutual dependence on pressure and flow rate in a pipeline system with different cross-sections

Hence, flow rate is higher inside a stenosis. Due to the sudden increase in the lumen after the stenosis, the corpuscular blood components are caught up in turbulences (detachment phenomena) with retrograde flow components (Fig. 5.1.1.4).

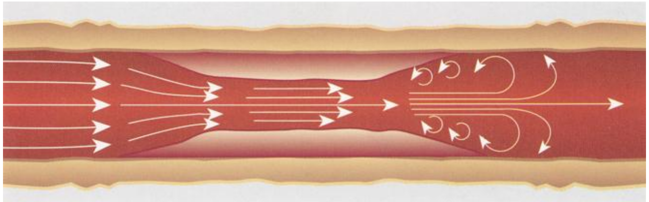

Fig. 5.1.1.4
Turbulences of corpuscular blood components due to detachment phenomena after the stenosis

A stenosis in the course of the carotids is therefore represented visually by a ragged curve due to the turbulences which is further away from the zero line. At the same time the acoustic, usually pure signal is disturbed and is represented as a high frequency hissing or crunching (walking in gravel).

After the stenotic area the signal either returns to normal (Fig. 5.1.1.5) or due to a high degree of stenosis may even show a reduced amplitude.

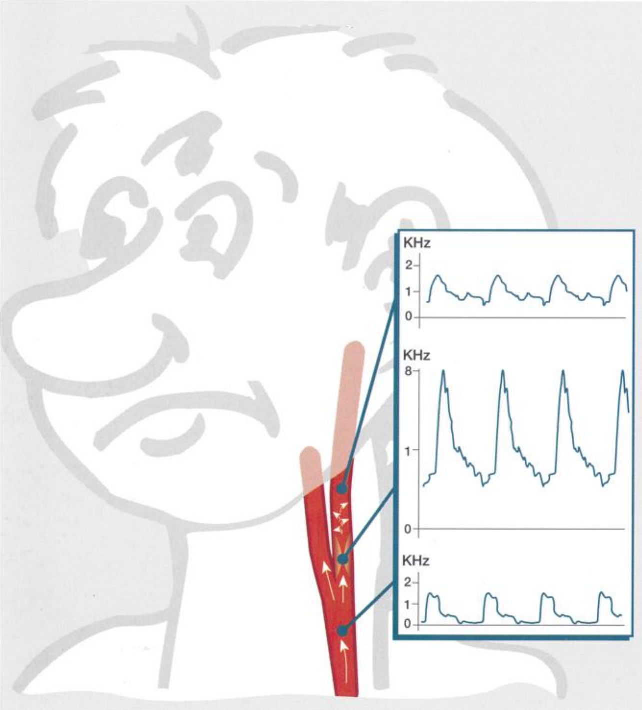

Fig. 5.1.1.5
Changes in the shape of the curve when passing through a carotid stenosis

As it can be difficult to differentiate between the individual carotid trunks in the case of pathological signals, a number of compression tests are used to clearly identify them. For example, to clearly identify the external carotid artery, it is recommended that the superficial temporal artery to the ventral side of the tragus is rhythmically compressed. This is expressed acoustically and visually by a recoil effect (Fig. 5.1.1.6).

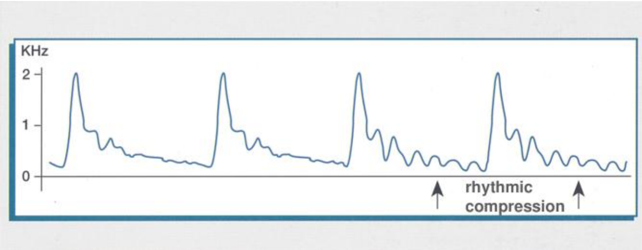

Fig. 5.1.1.6
Representation of the external carotid artery with recoil phenomena due to rhythmic compression of the superficial temporal artery

If the internal carotid artery is subjected to a Doppler investigation, the compression test has no effect on the Doppler signal.

Due to the bulbous type dilatation of the internal carotid artery in the area of the bifurcation, physiologically turbulences and zero-flow areas occur at this point which favour blood clotting (Fig. 5.1.1.7).

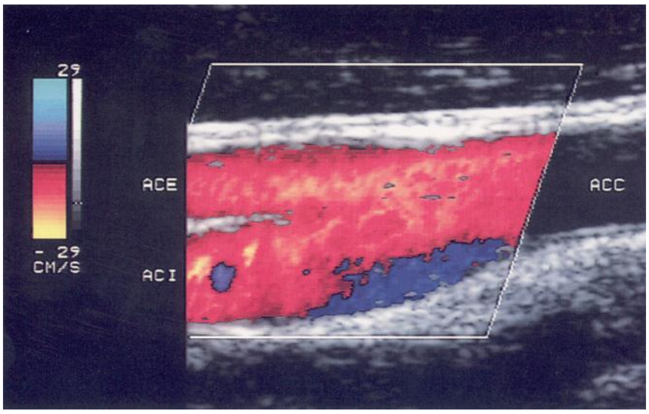

Fig. 5.1.1.7
Colour duplex representation of the physiological backward flow at the exit of the internal artery (blue)

Therefore, fatally, stenosing processes are frequently found in this area. In addition to a direct Doppler examination of the common carotid artery, the internal and external carotids, the collateral circulation is also examined via the ophthalmic terminal branches supratrochlear artery and supraorbital artery to obtain additional information.

Indirect Application of Sound to the Carotis Flow area 5.1.2.

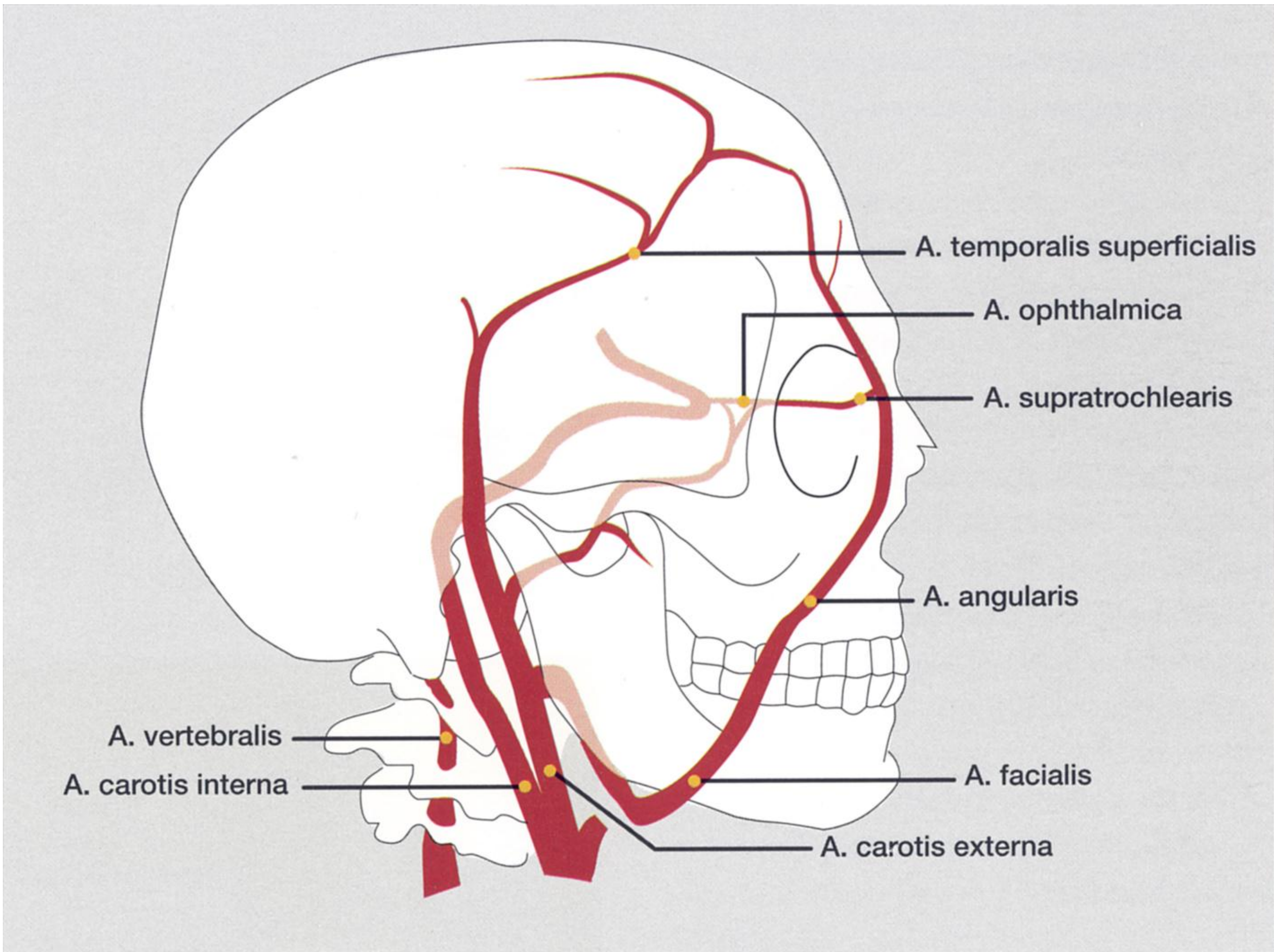

Fig. 5.1.2.1.
Representation of the ophthalmica collaterale

As shown in Fig. 5.1.2.1, intracranially, the ophthalmic artery originates at the level of the carotid siphon and creates a connection between the external and internal circulation (ophthalmica collaterale). As a subsidiary of the internal carotid artery, the ophthalmic artery and its terminal branches are usually supplied by the internal carotid artery. This means blood flows from the inside of the skull to the outside towards the eye. As a result, bidirectional devices register a forward directed flow when the supratrochlear artery in the corner of the eye is examined. In the case of a severe stenosis or an occlusion of the internal carotid artery, the ophthalmic artery is supplied retrograde by external circulation branches via its terminal branches of the supratrochlear artery and supraorbital artery due to the altered pressure gradient between the internal and external circulation. Now the Doppler device registers a backward facing flow (below zero). This way the external carotid artery takes over the supply of those parts of the brain normally supplied by the internal carotid artery thus preventing the loss of function of those parts.

As loops are frequently found in the periorbital vessels, evidence of a retrograde flow is not sufficient for diagnostic purposes as a retrograde flow can be feigned. For this reason the compression test of the superficial temporal artery and the facial artery should also be performed in this case. If the flow is truthfully a retrograde flow, the blood flow speed decreases significantly when both vessels are compressed. Blood flow can even come to a standstill or be turned around. The Doppler device registers a Doppler curve which drops down towards the zero line. (Fig. 5.1.2.2).

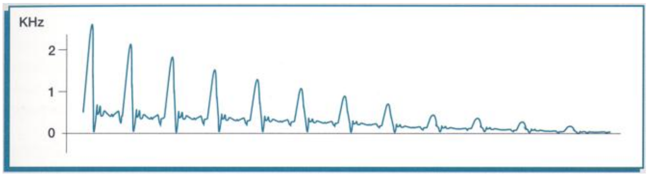

Fig. 5.1.2.2
Decrease in the blood flow in the supratrochlearis when compressing the temporal artery and the facial artery due to a stenosis of the internal carotid artery.

Application of Sound to the Vertebralis Subclavia System 5.1.3.

As directional Doppler sonography can be helpful in diagnosing a vertebral-basiliary insufficiency, the vertebral artery must also be examined with a Doppler device when examining the arteries supplying the brain.

Furthermore a proximal flow obstacle in the subclavian artery can have an effect on the circulation in the brain in terms of a „subclavian steal effect“ which is why this vessel should also be examined to complete the examination of the extracranial vessels.

Application of Sound to the vertebral artery 5.1.3.1.

It is usually best to examine the vertebral artery at the atlas loop. A 4 MHz probe is required for this examination and it is placed approx. 1 cm below and dorsally to the mastoid process and directed towards the contra-lateral zygomatic bone.

The signal of the vertebral artery has the flow characteristics of a cerebral vessel. As frequently the distal extracranial part of the internal carotid artery is picked up, the vertebral artery must be clearly identified. To do this, compress the common carotid artery on the same side for a short time, this should have no influence on the Doppler signal of the vertebral artery.

If, in the region of the atlas loop, a vessel with a low diastolic flow is picked up, it is usually the occipital artery. In contrast to the vertebral artery this can be compressed easily with the probe.

Depending on which branch of the atlas loop has been recorded, the flow of blood will either be towards the probe or away from the probe. Therefore it is usually not possible to assess the direction of flow. However, together with the examination of the subclavian artery it is possible to obtain sufficient information on the function of the vessel thanks to the characteristic flow of cerebral arteries.

If the vertebral artery is not found in the vicinity of the atlas loop, an examination at its exit at the “highest“ point of the subclavian artery should take place in the supraclavicular hollow. Here too the experienced examiner will find a signal typical of cerebral arteries which is physiologically directed towards the probe. As the thyroid gland vessels show similar flow behaviour, a differentiation must be made between these vessels and the vertebral artery. Clear identification of the vessels is possible using rhythmic compression of the atlas loop which is noticeable as a kick-back effect in the Doppler signal.

If the vertebral artery cannot be found here either, there is either an occlusion or a hypoplasy or aplasy (relatively frequent and of no pathological value) of the vessel. Further clarification could possibly be obtained using duplex sonography.

Application of Sound to the A. subclavia 5.1.3.2.

The subclavian artery is always examined in the supraclavicular hollow or infraclavicular further distally. Physiologically the signal is that of a high resistance vessel with a distinctive early diastolic backward and late diastolic forward flow. A pathological flow signal shown by a lack of the early diastolic backward flow component, is an indication of a haemodynamically relevant proximal flow obstruction. As a result, the vessel should be examined for any indication of the so-called subclavian steal effect via the vertebral artery (see that section).

The "Subclavian Steal Effect" 5.1.3.3.

In the case of the subclavian steal effect the arm concerned is supplied by the intracranial vessels, the circulus arteriosus willisii and finally the vertebral artery (Fig. 5.1.3.3.1).

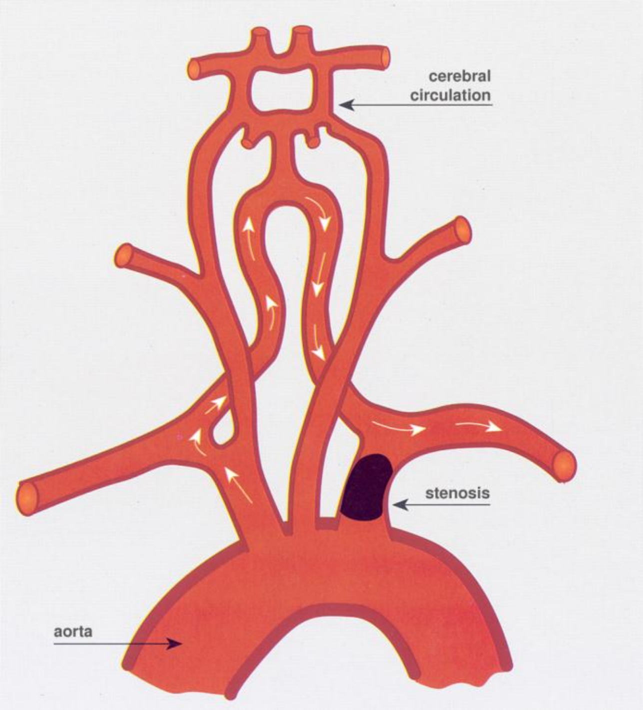

Fig. 5.1.3.3.1
Subclavian steal effect

In the case of a complete steal effect, flow is reversed in the vertebral artery at its exit. This means that the direction of flow is away from the probe. The vertebral flow can be normal to a great extent or can show an oscillating flow due to equal pressure between the vertebral artery and a possibly collateralized subclavian artery. In this case, to confirm the "subclavian steal effect", use a compression test on the homolateral arm. To do this pump up a blood pressure cuff to above systolic values whilst at the same time insonating the same sided vertebral artery. In the case of a subclavian steal effect this results in a reduction in the systolic and diastolic amplitude or in a reversal in the flow of blood in the vertebral artery. If this obstruction is loosened after 2 minutes, there is an increase in the blood flow in the direction of the arm due to the post-ischaemic hyperaemia .

Quantification of the Severity of stenoses using Bidirectional Doppler? 5.2.

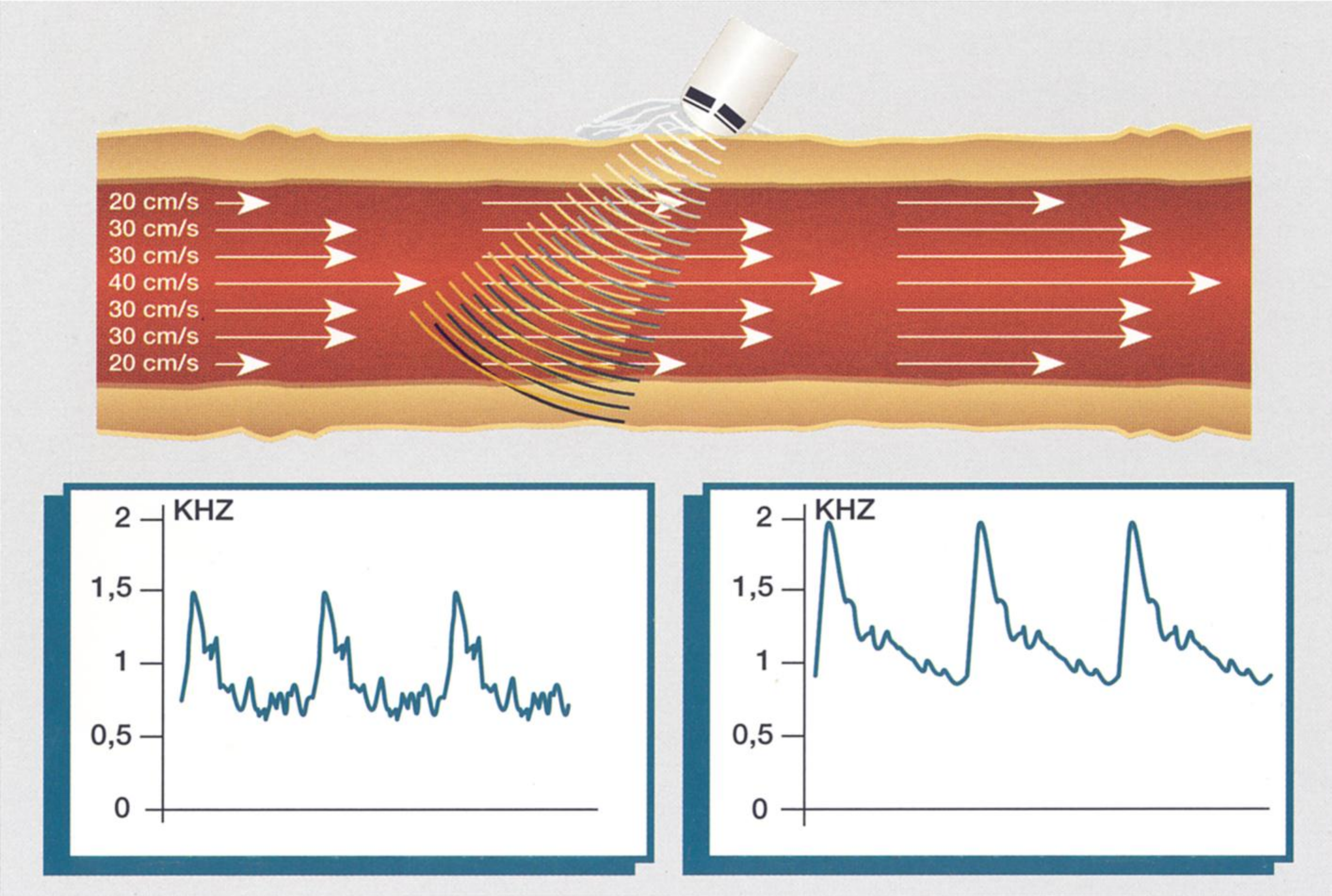

Fig. 5.2.1
Different degree of deflection in the case of laminar flow profile when using conventional methods of measurement (left) and frequency analysis (right)

Diagnosis using bidirectional Doppler devices without frequency spectrum analysis is limited to the qualitative assessment of vascular processes. However, a quantitative assessment based on the maximum frequency within a stenosis as quoted in numerous references is not permitted.

Explanation:

Conventional bidirectional Doppler devices (so-called Zero-Crossers) without frequency spectrum analysis ignore the fact that the corpuscular blood components move through the cross-sections of the vessels at different speeds. For example, in the physiological case, all the particles flowing at the edge of the vessel move slower than the erythrocytes in the centre of the vessel due to the friction resistance (Fig. 5.2.1). However, the Doppler flow curve recorded by the Doppler device can only represent one of these speeds. Contrary to general assumption, it is in fact not the maximum frequency that is recorded nor is it the speed averaged across the cross-section i.e. the average or mean frequency.

Instead, the Doppler flow curve represents the so-called modal frequency. This is the Doppler frequency which is caused by the most frequently occurring speed. In the example in Fig. 5.2.1 this would be 30 cm/s which is equivalent to 1560 Hz.

For a physiological laminar parabolic flow profile, the modal frequency is only approximately 70% of the maximum frequency. If flow is inhibited due to a stenosis, extremely high maximum frequencies are recorded in the area of the stenosis, but these are caused by only a very small number of erythrocytes. The majority of the corpuscular components

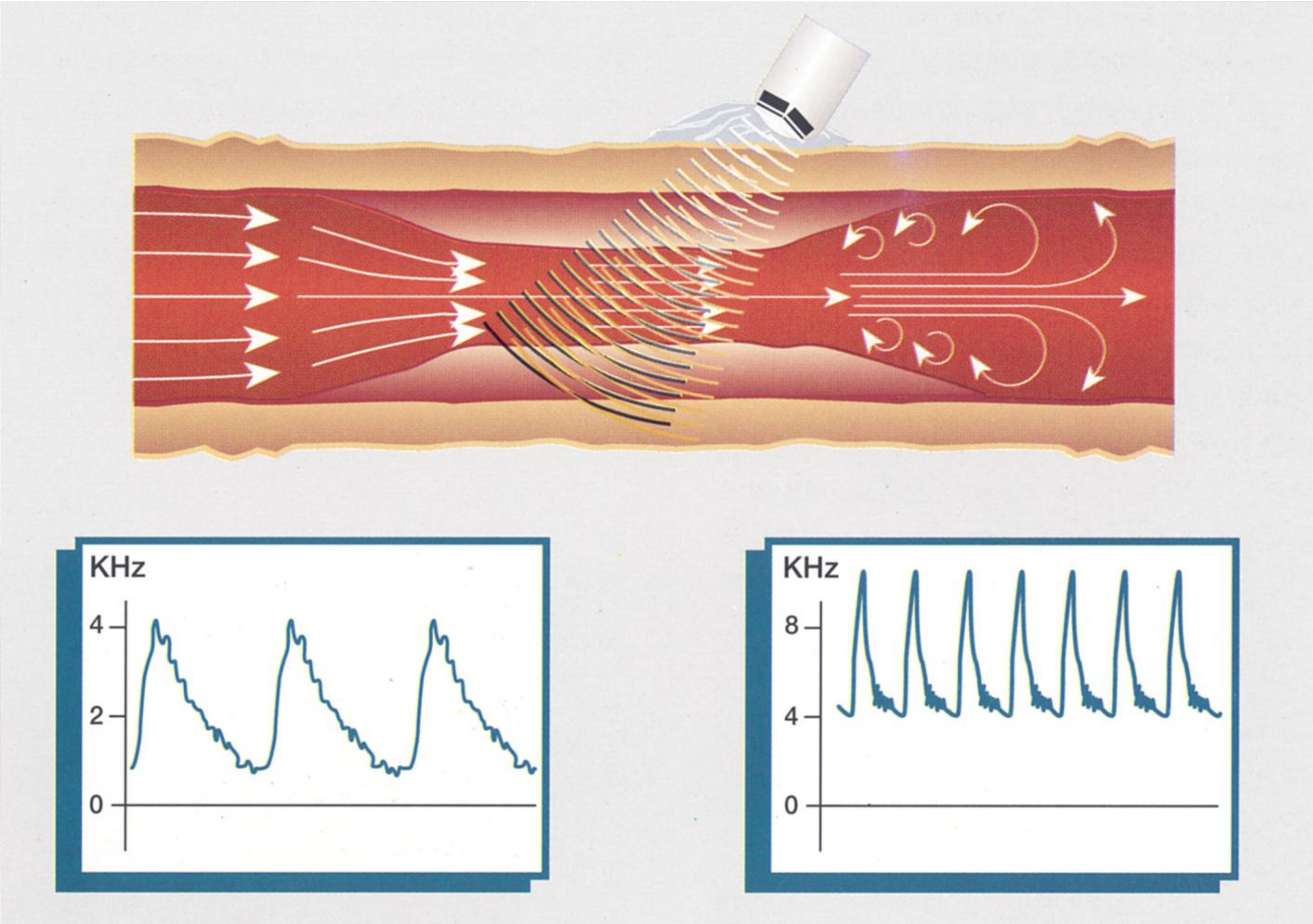

Fig. 5.2.2
Different deflection for inhibited flow profiles using conventional measuring technology (left) and frequency spectrum analysis (right)

move at a far lower relatively equal speed (Fig. 5.2.2). The modal frequency is now only a small proportion of the maximum values. For this reason the degree of deflection of these Doppler flow curves must not be compared with the maximum values specified in the reference literature which have been recorded using frequency spectrum analysis.

Diagnosis using Frequency Spectrum Analysis 5.3.

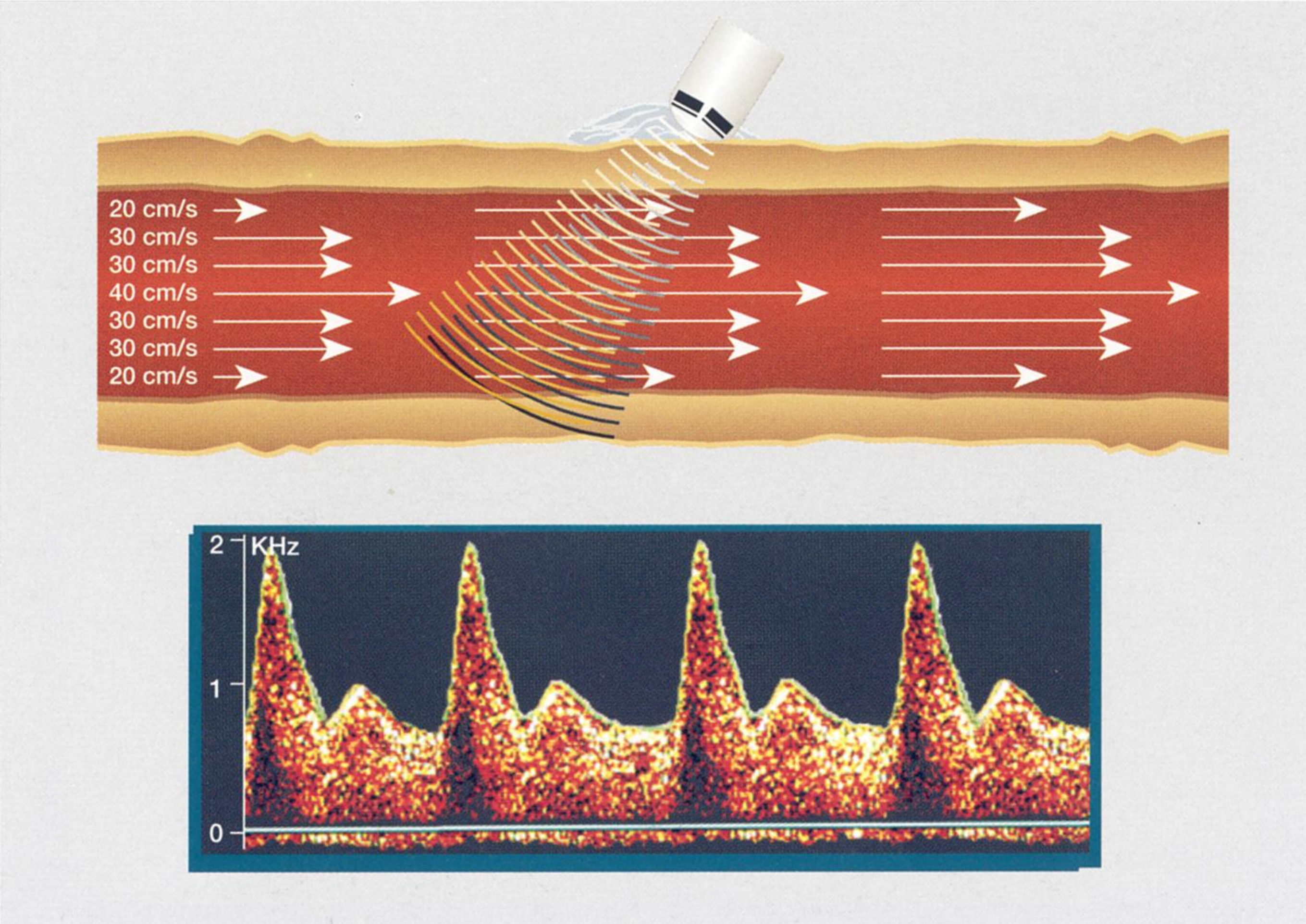

Fig. 5.3.1.
Representation of a narrow band spectrum

As described in section 2.3., frequency spectrum analysis takes into account all flow rates passing through the cross-section of a vessel. Representation of the spectrum facilitates the distinction between the flow profiles of the individual vessels and is impressive in documenting the interrupted haemodynamics. In addition, the maximum frequency measured with the help of this technique correlates with the degree of the stenosis.

In the case of a laminar parabolic flow the majority of particles – with the exception of the erythrocytes at the edges of the vessel – move with approximately the same speed in fixed parallel paths. In this case the result of the frequency spectrum analysis shows a narrow band spectrum (Fig. 5.3.1). Low speeds are under represented especially during systole which results in the so-called “systolic“ window. This is the dark area with only a small number of spots underneath the covering ends.

In the case of a stenosis, the erythrocytes leave their normal paths. Accelerations, turbulences, eddy currents (detachment phenomena) occur. In this case, a wide band spectrum is shown (Fig. 5.3.2).

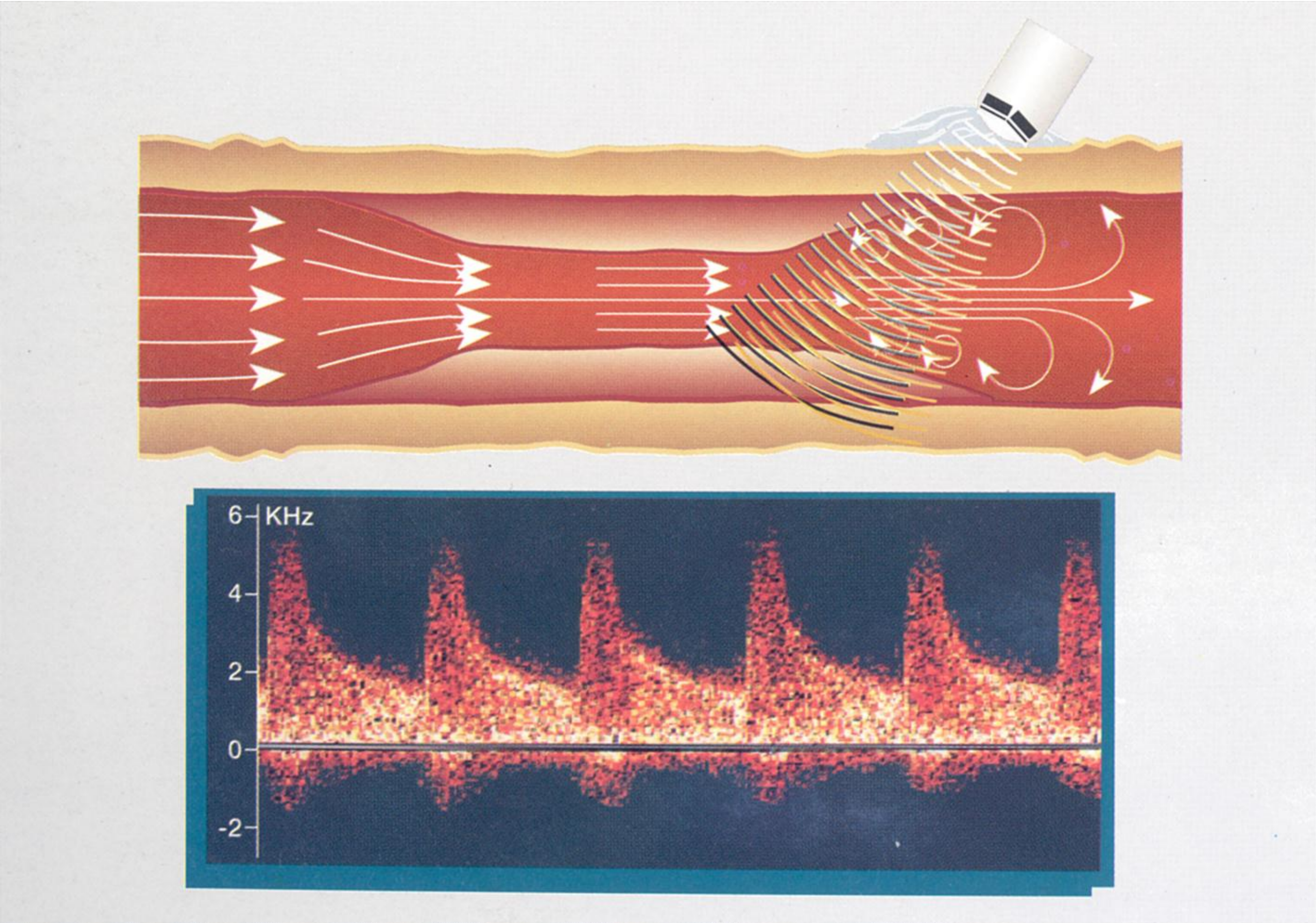

Fig. 5.3.2
Representation of a wide-band spectrum with retrograde flow parts

The top speeds are higher, the systolic window is missing and retrograde flows are shown as dense clouds pf spots underneath the zero line.

In addition, frequency spectrum analysis provides the possibility of including angle independent indices for the assessment of the degree of the stenosis, the most important ones have been described below:

Pulsatility Index

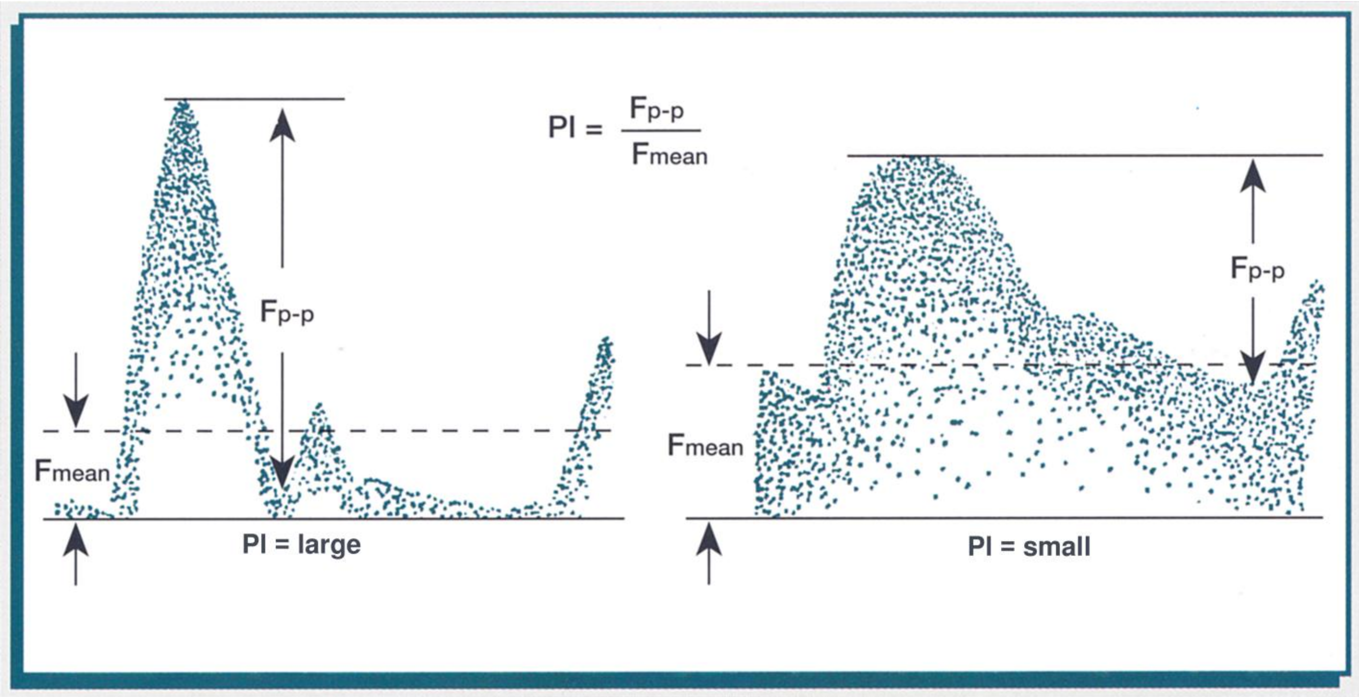

Fig. 5.3.3
Description of Doppler indices

To determine the pulsatility index according to Gosling, the quotient of the peak-to-peak-frequency-amplitude (Fp – p) and the average frequency averaged over time (TAM) is formed:

$$PI = \frac{FS - FM}{TAM}$$

The flow signal is determined distally, i.e. in the direction of flow after the stenosis, whereby the distance between the stenosis and the position of the probe is not important. On a muscle supplying vessel in which blood flow is normal, the PI is large (greater than 6) as the diastolic flow rate is low, and vice versa, the PI decreases if the difference between the systolic and the diastolic flow rate is low as is the case after a stenosis (Fig. 5.3.3).

In a healthy person the variance is considerable as the PI scatters between 6 and 13. A PI smaller than 6 indicates a small obstacle in the flow. A PI smaller than 4 indicates a considerable stenosis and a PI less than 3 indicates a severe stenosis or occlusion.

Resistance Index

The Resistance Index according to Pourcelot allows conclusions to be made concerning the flow resistance of an artery. The equation to calculate the RI is:

$$RI = \frac{FS - FD}{FS}$$

Here, FS is the systolic peak frequency and FD is the end diastolic peak frequency. When examining an artery which supplies a muscle in the extremities, values around 1 indicate a high vessel tonus, whereas values below this indicate post-stenotic vasodilatation.

S / D-Ratio

Fig.5.3.4
Comparison between amplitude heights of a spectrum representation with the maximum frequency curve determined using frequency spectrum analysis

The so-called S / D Ratio is also related to the Resistance Index. In this case the quotient is formed using the systolic peak frequency and the end-diastolic peak frequency.

$$SD-Ratio=\frac{FS}{FD}$$

As it is very complex to represent a frequency spectrum in terms of equipment, the purchase price of these systems is relatively high. On the other hand, they do overcome the limitations described of a zero-crosser and enable the examiner to make quantitative statements on the severity of a stenosis.

For this reason the latest developments are heading towards designing equipment which have an integrated frequency spectrum analysis unit but which do without the ability to display the spectrum.

Instead, the analyser calculates the real maximum frequency using the frequency spectrum determined, the height of the amplitude of which is equivalent to that of a spectrum and therefore also correlates with the severity of the stenosis. (Fig. 5.3.4).

Mean Frequency Curve

In addition, it also provides the possibility to represent the maximum and mean frequency at the same time. This means that the stenotic flow is represented impressively by other means as the greater the deviation of the mean frequency curve from the maximum frequency curve the greater the severity of the stenosis. (Fig. 5.3.5).

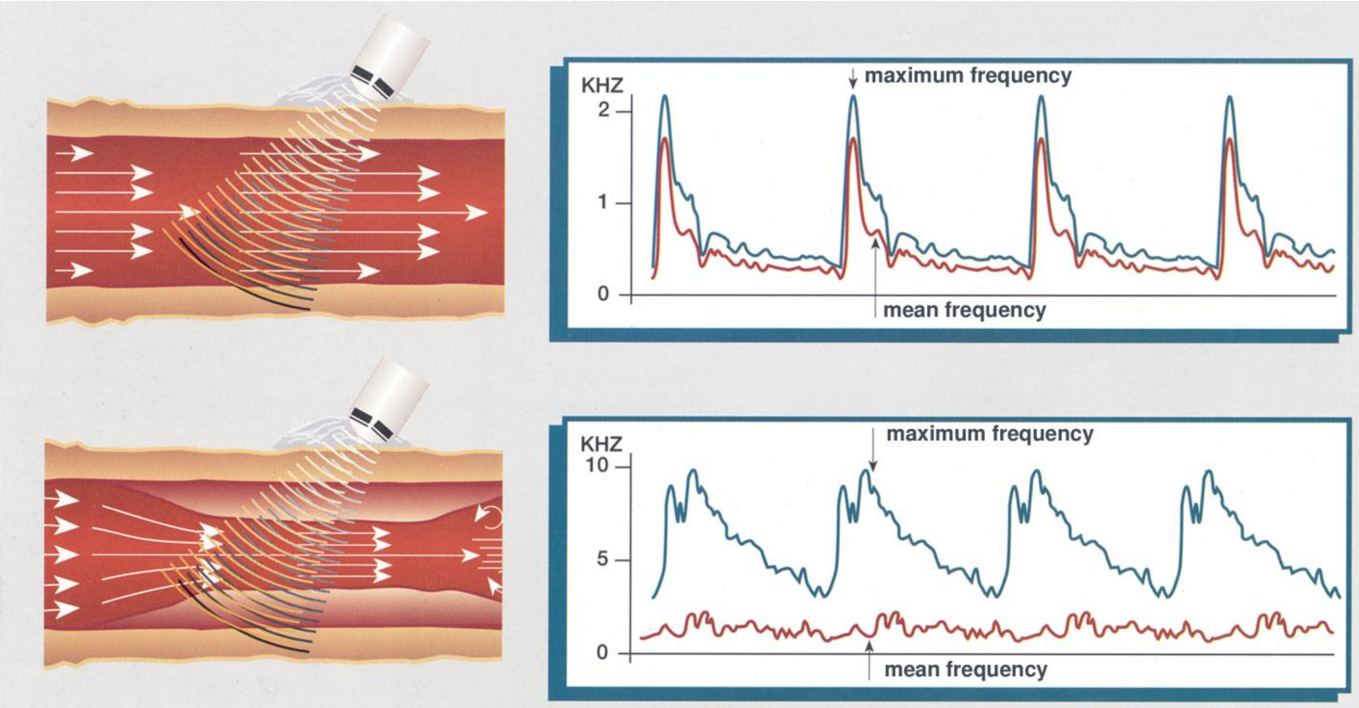

Fig. 5.3.5
Comparison between maximum and mean frequency curve in the case of uninterrupted and interrupted haemodynamics

As this equipment is virtually no more expensive than normal zero-crossers, but technically far superior, this type of device will soon force the conventional bidirectional Doppler off the market.

Transcranial Doppler Sonography 6.

The use of a pulsed Doppler is an absolute necessity for transcranial Doppler sonography to be able to differentiate between the large number of intracranial vessels. In addition, this device enables individual vessel sections to be assessed. As the sonic ray enters the vessel axis at an angle of virtually 0°, the Doppler shift is stated in cm/s instead of in kHz frequency shift due to the optimal angle of incidence. To insonate the bone, a low transmission frequency of approx. 2 MHz is used. To be able to penetrate the skull, higher transmission performance is used than are used for Continuous Wave Doppler devices. The transmission power can be varied on the device. It is recommended to start the examination with the lowest available transmission power. As the Doppler signals from intracranial vessels at greater depths are relatively weak, the conventional technology of a zero-crosser fails. For this reason the use of a frequency spectrum analysis device is absolutely imperative for this method of examination.

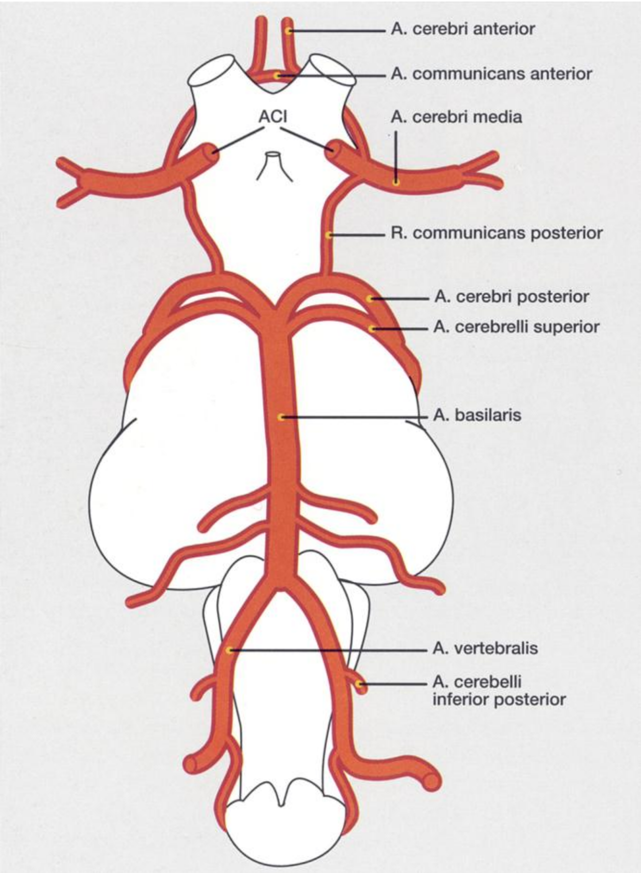

Fig. 6.1
Anatomy of the base of the brain (seen from below)

Literature describes three ways of insonating the intracranial arteries:

1. Transtemporal access to insonate the middle cerebral artery, anterior cerebral artery, posterior cerebral artery as well as the internal carotid artery.
2. Transorbital access to insonate the internal carotid artery (carotid siphon) and the ophthalmic artery as well as
3. Occipital access to insonate the vertebral artery and the basilar artery.

Fig. 6.1 shows the anatomical position of the vessels of the base of the brain (seen from below) with the arterial circle of Willis, whereby the development of this structure varies greatly. Complete development is only found in 20% of all cases (see Fig. 6.2).

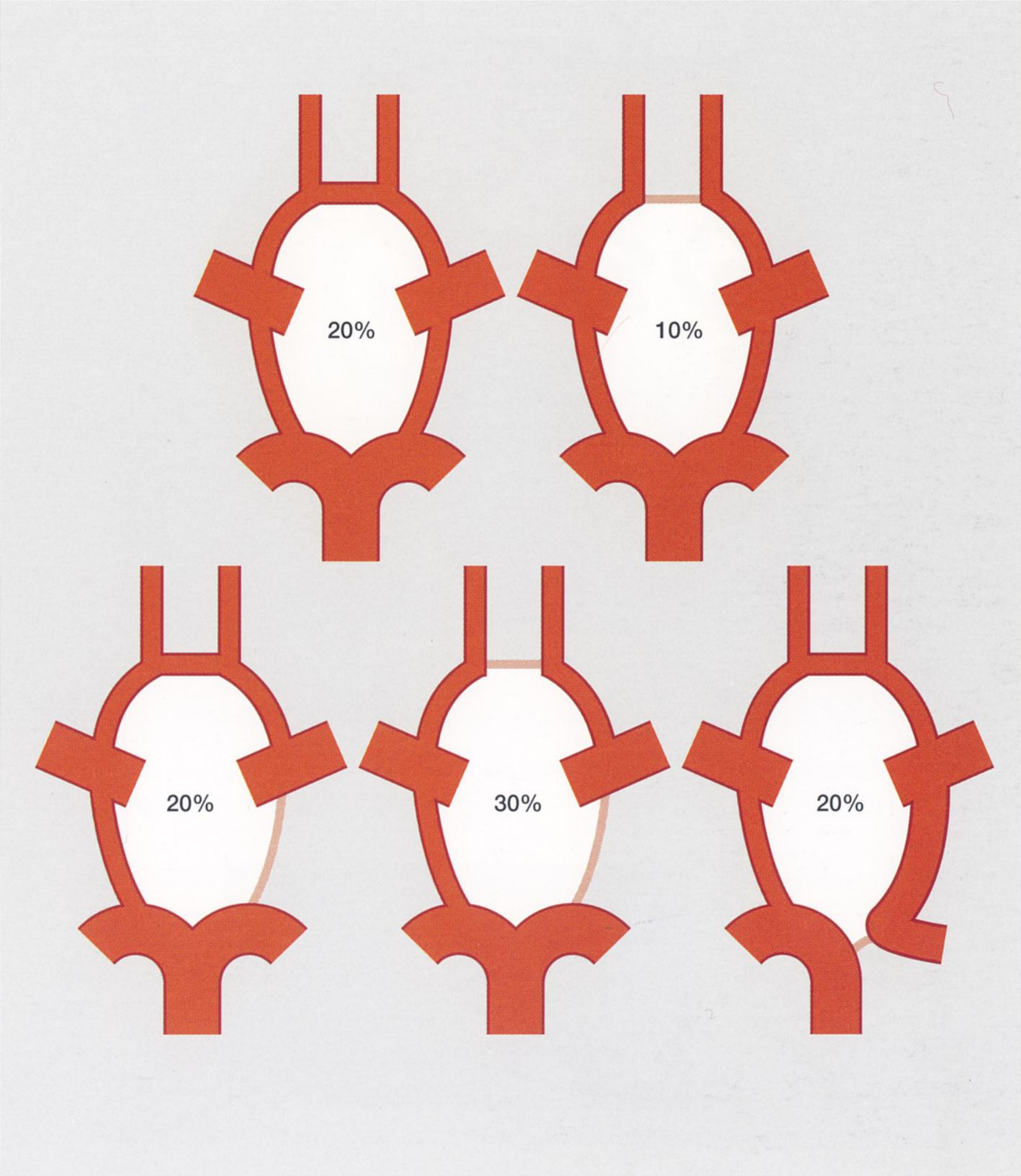

Fig. 6.2.2
Degree of development of the arterial circle of Willis and its frequency distribution (modified according to Widder)

Transtemporal access 6.1.

In the case of transtemporal access, a distinction is made between three different sonic windows (Fig. 6.1.1):

- one at the front in the area of the temple,
- one in the middle, in the area of the beginning of the upper pinna and
- one at the back above the pole of the pinna.

In most cases the examiner uses the middle access. This way representation of the middle cerebral artery is possible at a depth of approx. 50 mm which can then be followed back up to a depth of only 35 mm all the way to the periphery. However, searching for this vessel can take some time due to the differences in position in different persons. If an optimal signal is shown, you proceed to greater depths by shifting the measurement volume. In doing so, the position of the probe must be constantly corrected. At a depth of approx. 60-65 mm the examiner will find the internal carotid artery. The transition from the middle cerebral artery to the internal carotid artery is hardly noticeable.

Fig. 6.1.1
Different insonation windows of the transtemporal access

If the measurement volume is pushed even deeper, a signal directed away from the probe which comes from the anterior cerebral artery is formed in addition to the signals of the middle cerebral artery and the internal carotid artery which are directed towards the probe.

You are now in the area of the so-called T-fork. This T-fork can be recognized visually by the saw tooth pattern of resulting from the frequency spectrum analysis. The anterior cerebral artery can be followed up to a depth of approx. 75 mm where you then come across the anterior cerebral artery on the opposite side.

If the probe is shifted dorsally and aligned caudally, the posterior cerebral artery is found at a depth of approx. 55-65 mm. Here too the saw tooth pattern can occur in many cases if the opposite posterior cerebral artery is recorded at the same time at a depth of approx. 75 mm (Fig. 6.1.2).

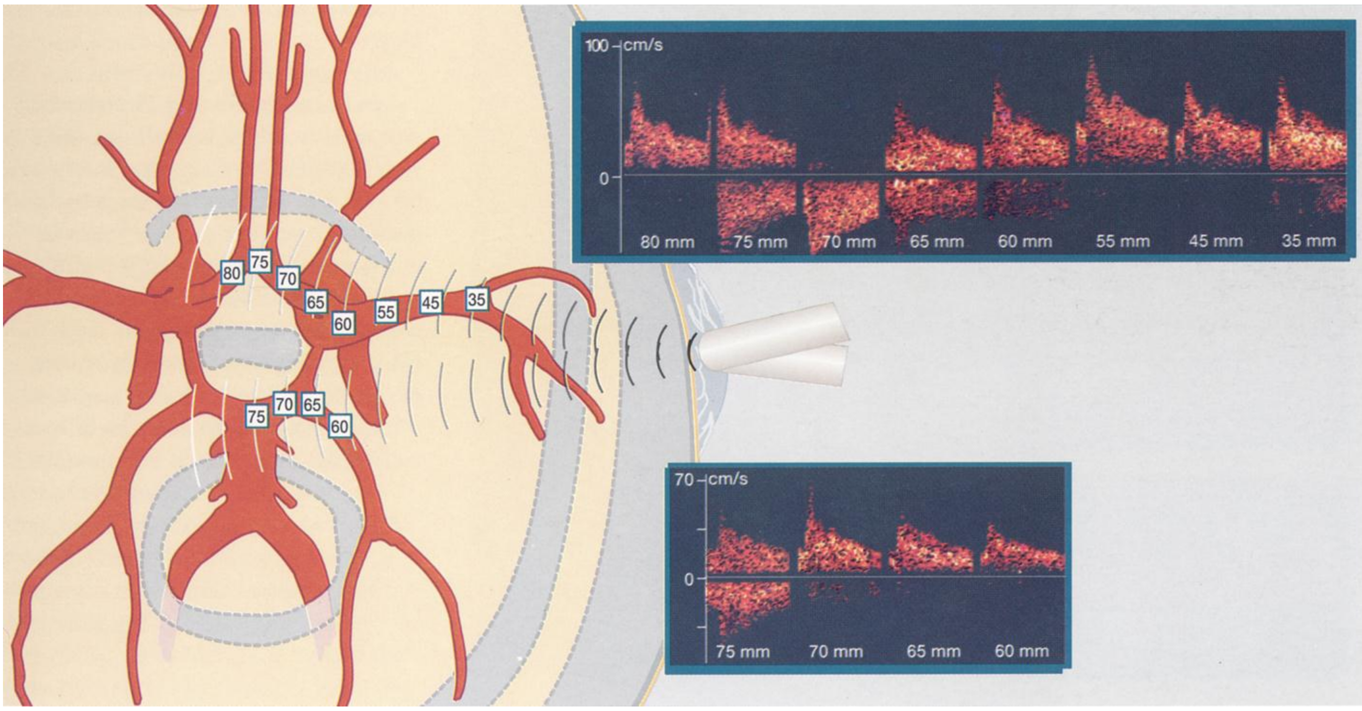

Fig. 6.1.2
Curve representation of the ACM, ACI, ACA and ACP at different depths of measurement

Transorbital access 6.2.

To insonate the ophthalmic artery and the internal carotid artery, the probe is positioned on the closed eye bulbus. To avoid damaging the lens, the transmission power of the device must be reduced as much as possible. Direct insonation of the lens should be avoided. Therefore the probe is to be positioned laterally to the lens.

The ophthalmic artery can be found at a depth of approx. 40-50 mm and it can be followed to the carotid siphon at a depth of approx. 60-70 mm. Depending on which branch of the siphon is being insonated, you will have a forward directed or a backward directed flow signal (Fig. 6.2.1). Frequently it is also possible to present the anterior cerebral artery via the transorbital access at a depth of approx. 70-80 mm.

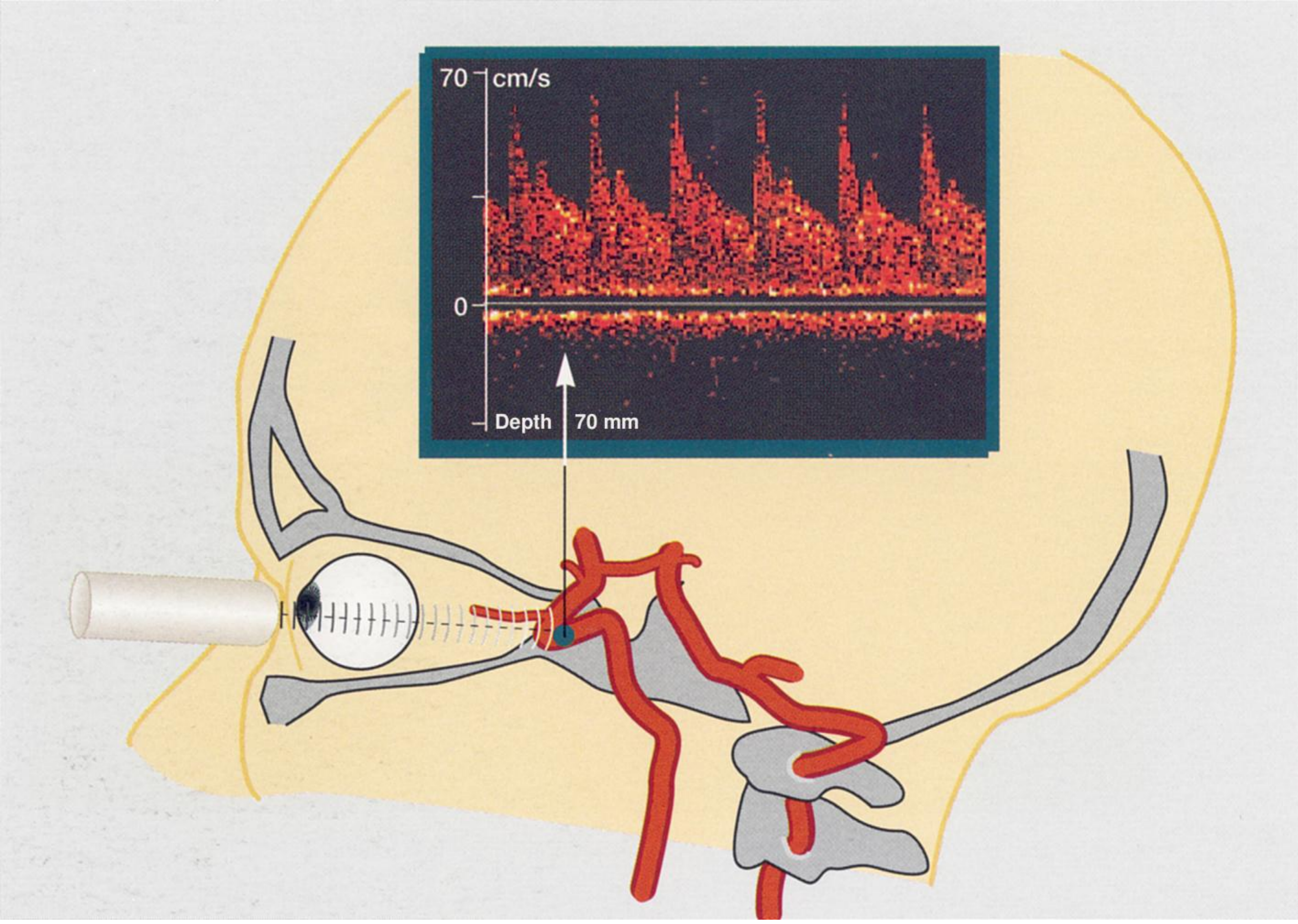

Fig.6.2.1
Insonation of the supplying branch of the carotid siphon via the transorbital access

The occipital access 6.3.

For insonation via the occipital access, the patient should be sitting. The patient's chin should be lowered towards his chest. The probe is positioned in the centre of the neck approx. 3 cm below the skull bone and the sound ray is pointed through the occipital foramen slightly towards the cranium. At a depth of approx. 60 mm the vertebral arteries can be found at slightly lateral alignment. At a depth of between approx. 90 and 115 mm the basilar artery is found. Both vessels show a direction of flow away from the probe.

Differentiation between the intracranial vessels and evidence of any possible vessel processes are frequently very difficult to find and frequently can only be achieved using compression methods on the common carotid arteries. As the effects of the compression test on the blood flow in the different sections of the vessels are extremely extensive, we have not provided them here to keep the size of this brochure reasonable. Please refer to the reference section at the end of this book for additional sources.

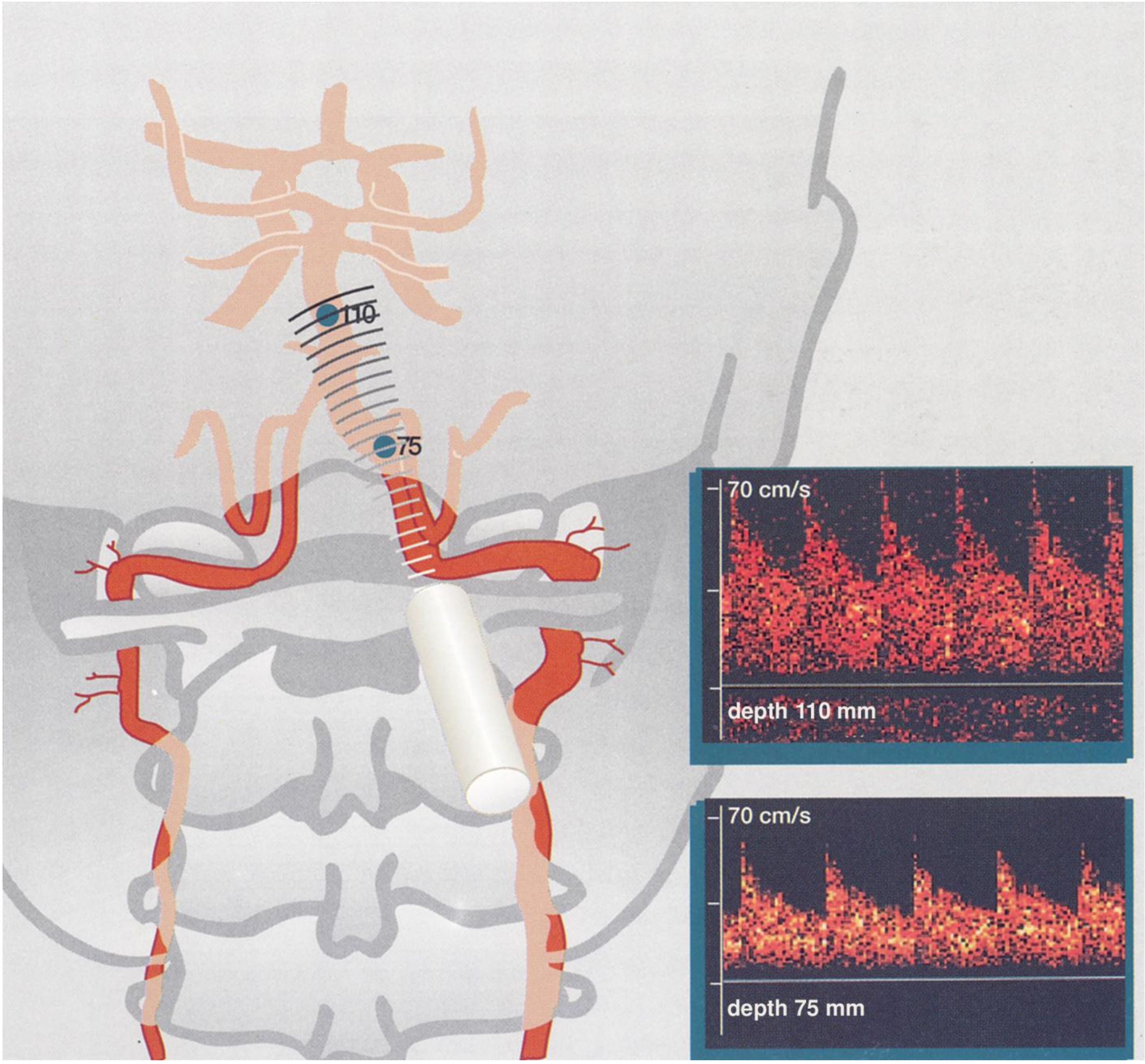

Fig. 6.3.1
Insonation of the vertebral artery and the basilar artery via the occipital access

Intracranial Doppler sonography is mainly suited to examine haemodynamic effects of extracranial vessel processes as well as to exclude intracranial stenoses before extracranial vessel surgery.

Veins 7.

Doppler sonography is ideally suited to diagnose the deep vein system of the extremities as well as that close to the surface. Sensitivity and specificity of the method in the diagnosis of thrombosis and the assessment of the valve function is specified at more than 90% - providing the examiner is sufficiently experienced.

In terms of technology both unidirectional and bidirectional Doppler devices are suitable. Whereas unidirectional devices are pre-destined for fast and reliable diagnosis during home visits and on intensive care wards, only bidirectional devices provide clear documentation.

When sonicating venous blood flow, a distinction is made between s-sounds and a-sounds.

s-sounds (spontaneous sounds) are those flow signals which can be recorded spontaneously under standard conditions without any provocative manoeuvres.

a-sounds (augmented sounds) are venous flow signals which can only be recorded using provocative manoeuvres (manual compression of soft tissue).

Normal and pathological findings: s-sounds 7.1.

As already described above, venous signals are characterized by a virtually constant, low frequency howling sound. In the case of deep abdominal breathing, the signal is subject to flow fluctuations which are synchronized with the breathing. In the case of deep inspiration, the inferior vena cava is compressed due to the high intra-abdominal pressure and therefore presents the venous flow in the extremities with a high degree of resistance.

This means that during inspiration, flow is slowed down which comes to a standstill in the end-inspiratory phase or if the Valsalva manoeuvre is used. During expiration the flow is accelerated due to the low resistance in the vena cava (Fig. 7.1.1). This type of breathing modulated flow pattern indicates that there is no haemodynamically effective flow obstacle proximally to the point of isonation.

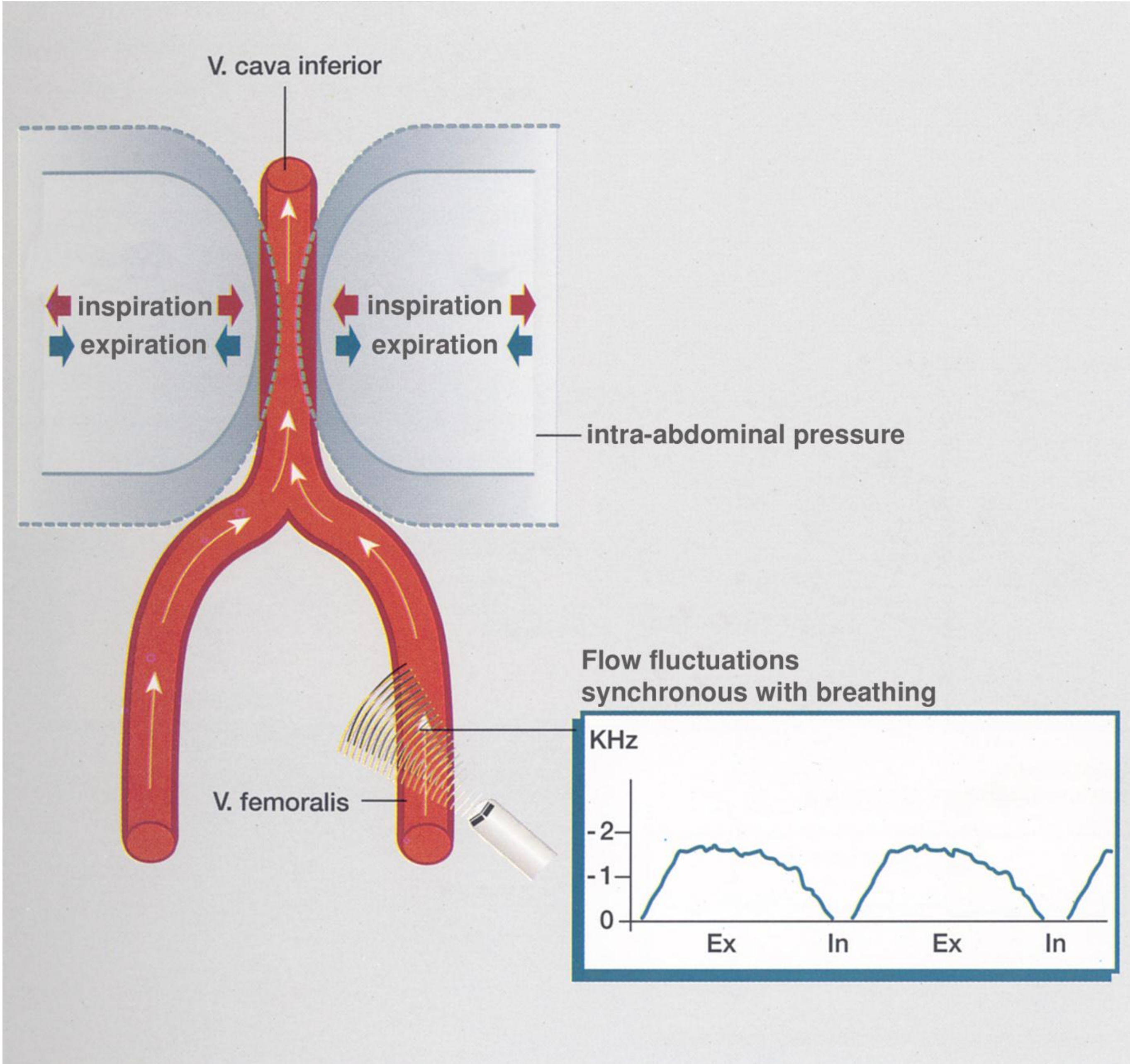

Fig. 7.1.1
Fluctuations in flow synchronous with breathing due to different compression pressure on the vena cava caused by changing intra-abdominal pressures

In the case of a stenosis, e.g. due to partial or complete formation of a thrombosis, the flow signal is different. Blood accumulates before the bottleneck. In the area of accumulation the pressure is greater than that in the intra-abdominal vessels. Hence blood will flow through the area of the stenosis at a constant rate, the flow which is synchronous with breathing is weakened or non-existent. (Fig. 7.1.2). In the case of valve insufficiency, venous blood may flow back at the end of inspiration.

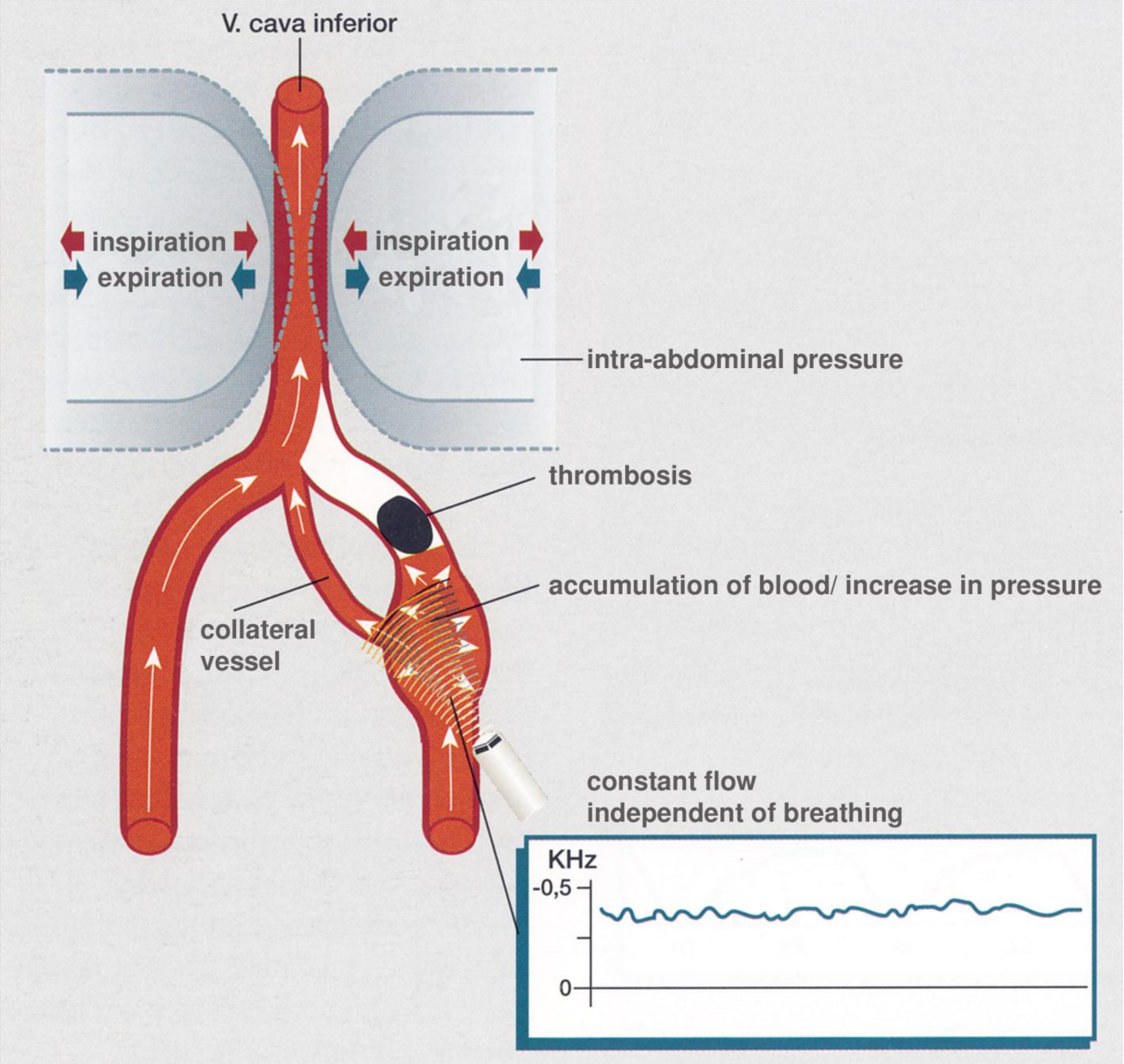

Fig. 7.1.2
Breathing independent flow signal in the case of thrombosis caused by pressure higher than intra-abdominal values in the area of blood accumulation

Normal and pathological findings: a-sounds 7.2.

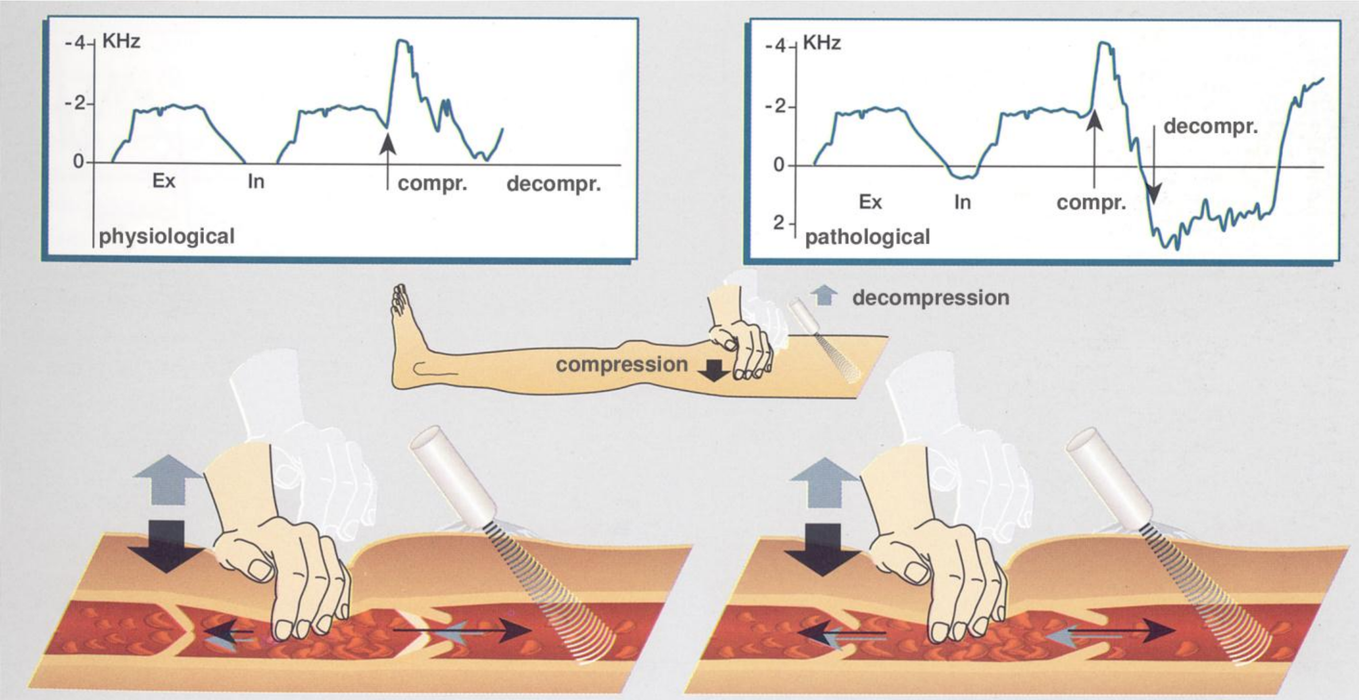

Fig. 7.2.1.
Changes in the flow curve caused by provocation manoeuvre distally to the area of examination

Additional information on whether a vessel is patent or not can be gained by compressing the soft tissue either distally or proximally to the area of examination.

In a healthy vessel, distal compression results in a rapid flow acceleration with high amplitude. During decompression no signal is received. If a flow signal is received during the decompression phase, there is a valve insufficiency in the vessel. In this case, the vessel fills up proximally due to the suction effect (Fig. 7.2.1).

In the case of complete obstruction in the immediate distal or proximal area of the probe, no a-sound is received. In the case of an incomplete shift of the vascular lumen it is weaker than usual.

In the case of sufficient valve conditions, proximal compression (or a Valsalva manoeuvre) results in a brief backward flow (max. 1 second) until the valves close. After this manoeuvre has been completed there is a greatly increased orthograde flow (“overshoot”).

In the case of valve insufficiency, a long retrograde signal is recorded during compression (Fig. 7.2.2).

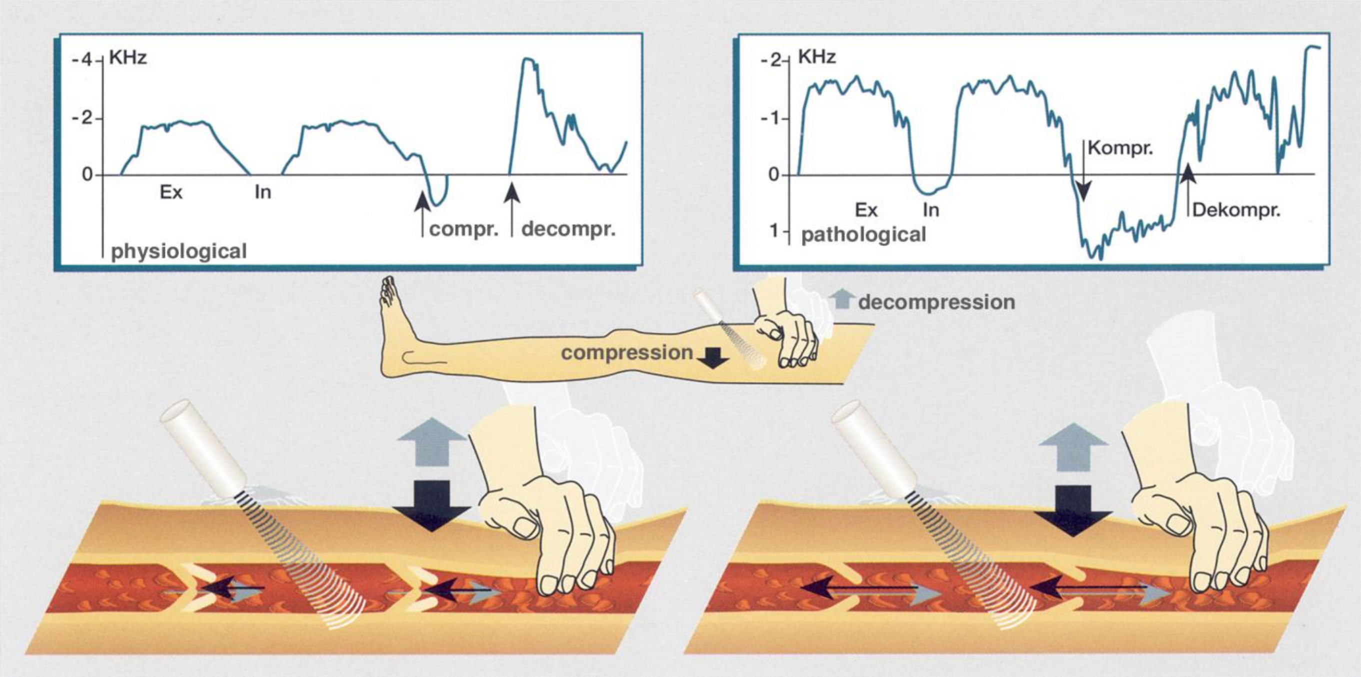

Fig. 7.2.2
Changes in the flow curve caused by provocation manoeuvre proximally to the area of examination

The examination process 7.3.

It is obligatory to examine the common femoral vein, the popliteal vein as well as the great saphenous vein.

To localize occlusions and insufficient valves in the deep thigh veins more accurately, it is recommended that the superficial femoral vein is also examined. Taking into account the scope of the brochure, we have not included a description of the examination of the small saphenous vein, the posterior tibial vein or the perforans veins as this is only relevant in special cases.

As the anatomy of each person is different and the characteristics of the blood flow also differ between people, but they do not differ within a person very much, assessment of the different areas of insonation is always made by comparing it directly with the opposite side of the same person.

At the beginning of the examination it is also recommended that the common femoral artery signal is found on both sides as an increased arterial flow of blood (e.g. due to inflammatory hyperaemia) can also influence the venous flow.

To find the deeper veins the examiner needs to find the reference arteries. For example, the middle common femoral vein and the superficial femoral vein are positioned laterally to the corresponding artery.

The popliteal vein is above the knee joint gap laterally from the artery, whereas below it, it crosses from dorsally to the middle. The great saphenous vein is not associated with any artery. The spontaneous flow signals (s-sounds) are evaluated, as is their dependency on breathing and their behaviour during compression manoeuvres (a-sounds).

To examine the common femoral vein, the superficial femoral vein and the great saphenous vein, the patient should be lying flat on his back. The popliteal vein is examined whilst the patient is lying face-down on his stomach. The calf is raised slightly to decompress the vein. To examine the deeper vein the 4 MHz probe should be used, to examine the surface veins, the 8MHz probe should be used.

Special findings of the common femoral vein 7.4.

Valve insufficiency

Proper, spontaneous flow behaviour can be seen. In the case of deep abdominal breathing, the examiner can find the end-inspiratory reflux. Distal compression generates an inconspicuous a-sound. A clear, retrograde flow signal is found if the Valsalva manoeuvre is performed and in the case of decompression of the thigh.

Complete occlusion

No spontaneous flow and no dependency on breathing is found in the case of an acute occlusion of the pelvic veins and / or the common femoral vein, provocation manoeuvres do not trigger any signals. If the occlusion is sub-acute or chronic, a collateral circulation has been formed. This is usually subcutaneous, suprapubic to the opposing side and has a considerably smaller vascular cross-section than the iliac vein.

An increased spontaneous blood flow is found which is not breathing modulated. Distal compression shows a weakened, wider signal, if the Valsalva manoeuvre is performed, the orthogradic flow is usually not interrupted. The spontaneous signal can usually be modulated by means of suprapubic compression.

Incomplete occlusion

In the case of an incomplete occlusion, spontaneous flow is increased and breathing modulation is greater or not as great depending on the degree of re-canalization. Due to the destruction of the vein valves, end-inspiratory reflux may occur. Distal compression results in a plumper signal than the signal on the healthy opposing side, the Valsalva manoeuvre shows a clear reflux (except in the case of very slightly opened vessel volume). Frequently a collateral circulation can be found as described in the section on “Complete occlusion”.

Special findings in the popliteal vein 7.5.

Valve insufficiency

A spontaneous flow signal is not obligatory in the popliteal vein. If the valves proximally of the popliteal vale are also insufficient, the findings are comparable with those of the common femoral vein in the case of valve insufficiency. If sufficient valves exist in the proximal, deeper thigh veins, the Valsalva manoeuvre is normal. However, decompression after compression of the calf and compression of the thigh in the middle or distal third results in a retrograde flow signal.

Complete and incomplete occlusion

In principle the same criteria apply here as they do for the common femoral vein. The great saphenous vein provides valuable additional information. If the deep thigh veins or the popliteal vein have been displaced to be haemodynamically effective, the great saphenous vein serves as the collateral circulation and shows an increased blood flow.

A weakened signal via the popliteal vein in the case of distal compression without evidence of insufficient valves and with an increased blood flow in the saphenous vein indicates a possible deep calf vein thrombosis.

Special findings for the great saphenous vein 7.6.

Valve insufficiency

It is unusual to receive a spontaneous signal from the great saphenous vein. The examiner can locate it by passing the probe over the probable course of the vein without compressing the skin whilst using the other hand to compress the medial side of the knee joint. The whole length of the saphenous vein can be examined once it has been found. The criteria described above apply to assess valve conditions. An increased flow on one side compared to the other side always indicates a possible deep leg vein thrombosis.

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