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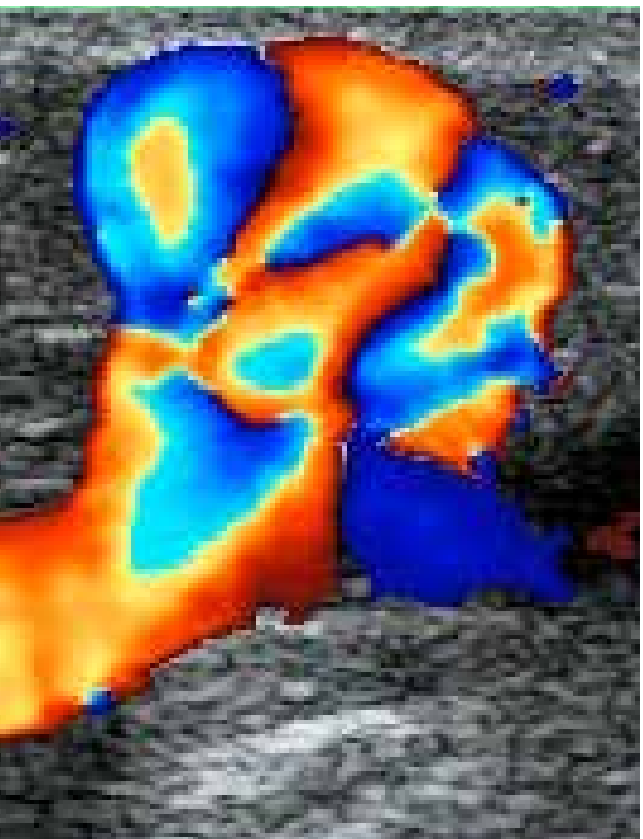
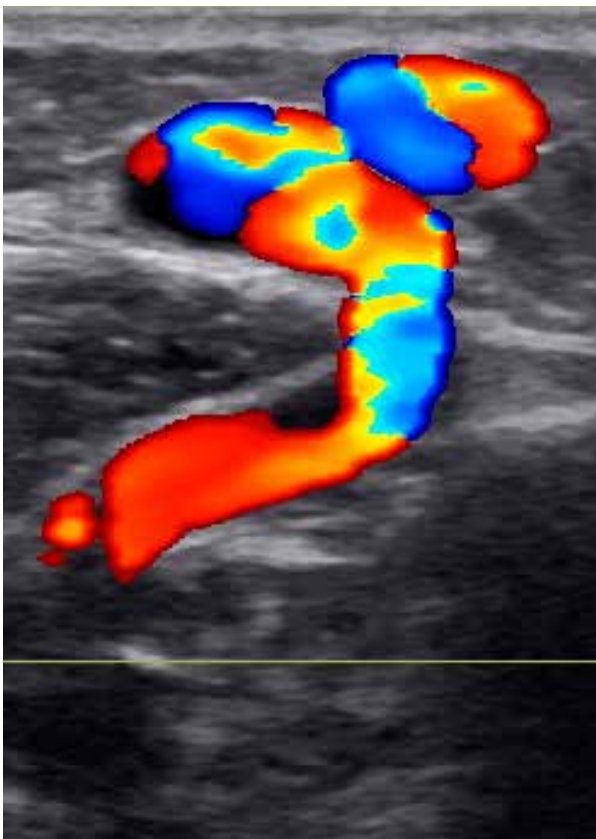
Ultrasound examination of the lower limbs

2nd edition, revised and updated





Dedicated to my teacher and outstanding physician
Professor MUDr. Ivo Krč, DrSc.

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Doc. MUDr. Dalibor Musil, Ph.D., and his colleagues

ULTRASOUND EXAMINATION OF THE LOWER LIMBS

2nd edition, revised and updated

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Contents

List of abbreviations

2D, 3D	two-Dimensional, three-Dimensional
a., *aa.*	artery, arteries
CAI	Color Amplitude Imaging
CD	Continuous Doppler
CEAP	Clinical, Etiological, Anatomical, Pathophysiological Classification
CFM	Color Flow Mapping
CPA	Color Power Angio
CW	Continuous Wave
DIP	Distal Insufficient Point
DVT	Deep Vein Thrombosis
GSV	Great Saphenous Vein
IUA	International Union of Angiology
IUP	International Union of Phlebology
LLL	Left Lower Limb
LL	Lower Limb
m., *mm.*	*musculus*, *musculi*
n., *nn.*	*nervus*, *nervi*
PD	Pulsed Doppler
PDGF	Platelet Derived Growth Factor
PE	Pulmonary Embolism
PIP	Proximal Insufficient Point
PRF	Pulse Repetition Frequency
PW	Pulsed Wave
RR	Relative Risk
RVI	Reflux Volume Index
SFJ	Saphenofemoral Junction
SPJ	Saphenopopliteal Junction
SSV	Small Saphenous Vein
st. p.	*status post*
TG	Total Gain
TGC	Time Gain Compensation
UGS	Ultrasound-Guided Sclerotherapy
US	UltraSound
v., *vv.*	*vena*, *venae*
VEGF	Vascular Endothelial Growth Factor
$V_{max.}$	Maximum rate of return vein flow
VSP	*vena saphena parva*
VSM	*vena saphena magna*

Preface

Seven years have elapsed since the first edition of this monograph which at the time, was given a very positive reception by the professional public. One of the chief reasons for this was that the book filled a large gap in the Czech market on the theory and practice of ultrasound diagnosis in venous disorders of the lower limbs. Appreciation of the first edition prompted us to write a second, revised and updated version.. Guiding our efforts in this were the discussions with doctors at congresses and the many fruitful suggestions gained from sonography workshops.

Routine use of ultrasound is revolutionising medicine in the diagnosis and treatment of a large number of diseases and conditions. In phlebology, it has become crucial. When carried out properly, ultrasound simplifies and makes everything easier for both doctors and patients. It enables fast and reliable diagnosis of superficial and deep vein thrombosis, identifies primary reflux sites in chronic venous disease and allows long-term follow-up of patients with vein disorders. For vascular surgeons, ultrasound mapping of the superficial veins should be indispensable before each varicose vein surgery.

Confronted with everyday practice, with the specific requirements and questions which doctors have when they refer their patients for ultrasound examination of the lower limbs, we have made efforts to develop the techniques that suit our particular clinical conditions and systematically expand our knowledge from the medical literature and lectures.

Following the first edition, this book updates the view of vein disease and the current position of ultrasonography in phlebology. It aims to be a useful text on ultrasound, a hands-on guide for new doctors and those in routine practice who are looking for answers to professional questions arising from their work. The publication is thus dedicated not only to doctors in the diagnosis of vein diseases, i.e. angiologists and radiologists, but also to the specialists who most often send their patients for scanning--surgeons, dermatologists, internists, cardiologists and general practitioners.

Each chapter is written as an independent unit and for this reason a certain amount of redundancy is unavoidable. Parts may be overly simple to readers familiar with ultrasound principles. It would be untrue however, to assume that all those who come in contact professionally with ultrasound, understand the technology.

As an imaging mode, an integral part of the book is drawings and images taken during an ultrasound examination and presented to the reader as they are seen in common practice. We hope you find this monograph useful.

doc. MUDr. Dalibor Musil, Ph.D.
Olomouc, 15th September 2015

1 Ultrasound in phlebology

Dalibor Musil, Ivo Hofírek

1.1 Technical Principles of Examination

1.1.1 What is ultrasound

Ultrasound (US) refers to high frequency mechanical oscillations transmitted to particles in a medium. If we imagine the waves as the concentric circles that expand on the surface of water following the impact of a stone, the circles are the regions of pressure change as from gas to liquid to solid. Ultrasound waves have a frequency greater than 20 kHz (20,000 Hz). This creates vibrations inaccessible to the human ear. **Production of these higher frequencies is due to piezoelectric crystals in the ultrasound probe** which have the ability to convert electrical impulses into mechanical vibrations, transmitted in pulses to the body in relation to their frequency. **For ultrasound diagnosis in medicine, the frequency band 2 MHz to 50 MHz is used, in common practice 3 MHz to 10 MHz.**

The sound is always propagated as longitudinal waves – the waves are parallel to the direction of propagation. In solids, the sound spreads not only as longitudinal but also transverse waves when the beam is perpendicular to the direction of the propagation. Ultrasound vibrations in media such as the soft tissues and fluids of the human body, spread along the longitudinal wave. Transverse waves only spread in bones. With increasing frequency, at high and very high frequencies (in the order of MHz), ultrasound waves behave like electromagnetic waves. The average rate of ultrasound propagation in the human body is 1540 m/s. The velocity is independent of the frequency used but depends on **acoustic impedance,** that is, the **resistance** that a US beam encounters in the tissue receiving it. Acoustic impedance is the characteristic that determines the relationship of US waves to the surroundings in which the ultrasound spreads. It is defined as the density of a substance multiplied by the sound velocity in the material ($kg\ m^{-2}\ s^{-1}$). An analogy is the optical refractive index. The acoustic impedance is proportional to the elasticity and density of the tissue and increases in the order of: lung – blood – soft tissues (internal organs) – muscle, bone (**Table 1.1**).

An **acoustic interface** (e.g. fat/muscle, bone/muscle, muscle/blood, etc.) is created at the point of contact of two media with different sound propagation properties. At this interface, the ultrasound waves partly reflect and partly transmit, provided the beam is perpendicular to the medium. If it is oblique, the waves are partly reflected and partly dispersed (**Figure 1.1**). Ultrasound waves are also absorbed by the tissues and acoustic energy is transformed into thermal energy (heating the tissues). Higher frequency US waves are absorbed the most. The US signal at an acoustic interface changes its energy (amplitude). The ratio of the amplitude of the transmitted and reflected ultrasound signal is called the **reflection coefficient** and is dependent on the acoustic impedance of the media (**Table 1.1**). Dispersion is the propagation of US waves into space in all directions. It occurs when the acoustic interface is less than the wavelength of the incident wave (e.g. erythrocyte dispersion). Reflected back to the probe,

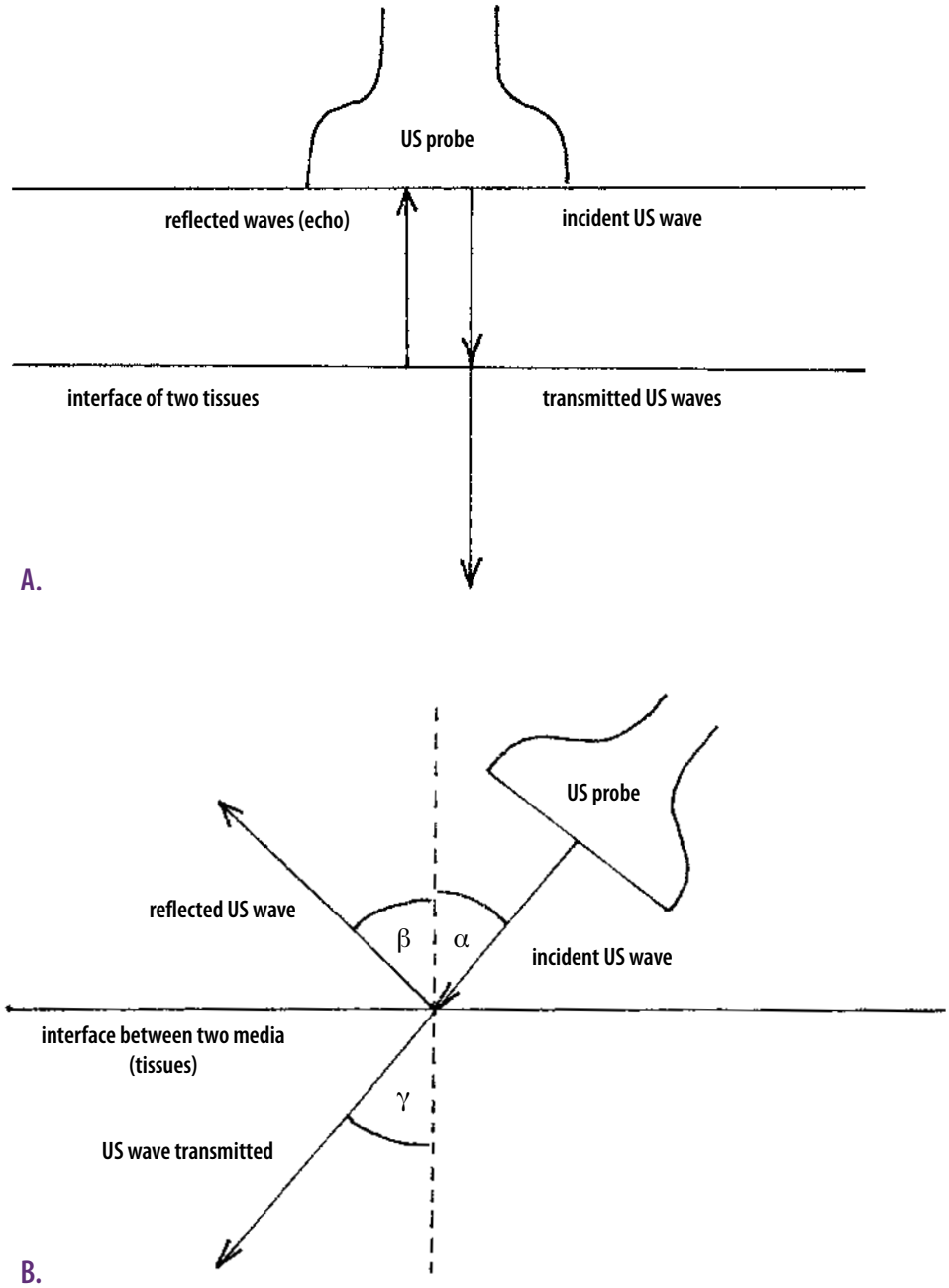

Fig. 1.1 *US wave propagation – At the tissue interface, a percentage of US waves reflect at the same angle but in the opposite direction to the incident wave, non-reflected waves pass through the second medium in the direction of impact (α = β = γ) –* ***A.*** *acoustic waves perpendicular to the interface of two media,* ***B.*** *acoustic waves incident obliquely at the interface of two media at an angle α*

is a US signal (echo) with a certain frequency, intensity and time delay. **The crystals in the probe, convert the US waves into electrical impulses**.

The ratio of the intensity of the reflected wave to the incident wave:

$$R = \frac{(Z_2 - Z_1)^2}{(Z_2 + Z_1)^2}$$

The greater R is, the greater the degree of reflection, i.e. R for a soft tissue interface such as liver and kidney is 0.01, i.e. only 1 % of the sound is reflected. For muscle/bone interface, 40 % is reflected and for a soft tissue/air interface 99 % is reflected.

Tab. 1.1 *Acoustic impedance for different tissues*

Medium	**Impedance** (kg m^{-2} s^{-1})
air	$0{,}000\,4 \times 10^6$
lung	$0{,}46 \times 10^6$
blood	$1{,}61 \times 10^6$
soft tissues	$1{,}63 \times 10^6$
muscle	$1{,}70 \times 10^6$
bone	$7{,}80 \times 10^6$

1.1.2 Creation of an ultrasound image

The basic and simplest type of ultrasound image is a one-dimensional recording of time sequence and magnitude (intensity) of the acoustic energy of reflections (echo) of the US signals transmitted to a tissue. This view is called the **A mode** (Eng. amplitude). Moving structures can be displayed in a continuous A view, which is **called M** (Eng. motion). In angiology and phlebology, the M image is not used being mainly the domain of cardiology.

A more advanced display type uses the brightness changes of individual screen dots emitted by incoming echoes in a range of up to 256-degrees-gray scale. This is referred to as the **B-mode, brightness modulation** and it produces a 2D image. The **dynamic B-mode** enables rapid simultaneous emission of signal and echo processing.

Ultrasound machines used in phlebology usually work in pulse mode (pulse Doppler, PW Doppler, see further). The probe sends a short pulse, an ultrasound signal of a certain frequency and the reflected echo is converted into electrical impulses that are processed as a 2D image. Each point on the monitor corresponds to the intensity of the received US signal – a specific brightness intensity in the gray scale from white to black. If pulses are transmitted parallel to received echoes, we refer to a **linear view**. The US image is rectangular. If the transmitted pulses and echoes are divergent, this is a **sector view** and the US image is diverging from the probe.

1.1.3 Processing ultrasound signals

We can simplify the capture and processing of US images if we imagine processing the images from a digital camera. The quality of the image depends on the conditions of taking the first shot ("photo"), adjusting the camera settings to obtain the type of picture desired (for example, sports, landscape, night scenery, etc.), the subject, lighting and a number of other parameters (**preprocessing**), chip resolution and chip size, data processing and noise reduction programs. In this way, the data is edited with specialised computer software (**postprocessing** – editing of the image).

Today, digital technologies allow for a number of adjustments to the final US image. These conform to the programs and technologies used in various types of US machines which of course make comparison difficult and create variation in the appearance of the same tissue.

Other settings that can edit and optimise the image, are **gain** (signal gain) and **compression** (signal condensation). These work differently in 2D display and color flow mapping (CFM) (see further). Gain and compression may also be default settings

US signals on absorption (change in acoustic energy through heat) are weakened, especially if they come from a greater depth. For this reason, the reflected echo from the human body must be sufficiently amplified called gain. Gain in US signal is achieved in three ways:

1. **Total gain (all incoming US signals are amplified) – using the total gain function (TG – Total Gain),** which is referred to in US machines as gain or 2D gain, and will amplify the image.
2. **Selective gain (gain is greater for US signals coming from greater body depth, i.e. later)** – using the **compensation time gain (TGC – Time Gain Compensation)** which is regulated in some US devices by slider keys, each key regulating gain from a certain tissue depth.
3. **Active amplification (amplifies the US signal coming from the pulse Doppler record).**

The dynamic range expresses the ratio between the strongest and weakest measurable echo in decibels (dB). This is important for image quality. Another image improvement is **harmonic imaging** which was originally used in cases of technically difficult US exams but it has found application in all other areas of US investigation. Instead of increasing the US power and extending the investigation time, a strong signal of a given frequency is transmitted to the area displayed to obtain a satisfactory image. The probe recursively retrieves the **natural harmonic waves with the 2nd harmonic.** Harmonics occur spontaneously in tissues due to the non-linear propagation of US waves. However, they are weak and need a powerful scanner to capture and appropriate software. The natural harmonic display shortens the scanning time, increases the contrast during routine screening and allows for better imaging in technically difficult to investigate patients. One caveat however, is that a better 2D image with harmonic display can compromise color mapping and Doppler measurements.

1.1.4 Doppler effect

For quantitative and qualitative blood flow investigation, the **Doppler effect** is indispensable. This phenomenon, first described by Austrian physicist, **Christian Doppler**

in 1842 while he was in Prague, was used in medicine for measuring the velocity of blood for the first time in 1960 by the Japanese, Satomura.

The Doppler effect is a physical phenomenon whereby the wavelength, electromagnetic or mechanical (sound, ultrasound) transmitted by a source is perceived by an observer as increased or decreased, if the source (transmitter) and the observer (receiver) change distance. Between the moving transmitter and the stationary or moving receiver, the acoustic signal is subject to a frequency shift, to a lower or higher frequency depending on whether the transmitter is away from (lower frequency) or towards the observer (higher frequency).

The probe transmits US waves to the body in constant frequency pulses. For immobile objects, US reflects without changing frequency.

The US waves picked up by the probe, are processed as the image (B-mode). from moving structures (e.g. erythrocytes), US waves are reflected as frequency. The difference between the frequency of transmitted and received waves, is called the frequency shift. The amount of frequency shift is proportional to the velocity of erythrocytes (blood flow rates). This relationship is expressed by the Doppler equation:

$$v = \frac{F_d \cdot c}{2F_v \cdot \cos\theta}$$

v – blood flow velocity
F_v, F_d – frequency of emitted (v) and incident (d) US waves
c – a constant indicating the rate of US propagation in the blood (1540 m/s)
θ (theta) – the angle between the direction of impact of the US wave and the direction of the erythrocyte movement
v – velocity of erythrocytes

If the angle of incidence is 90°, the Doppler equation is 0 (cos 90° = 0) and blood velocity measurement is impossible. To measure absolute speeds, it is advisable to select the smallest incidence angle. At an angle of up to 10°, the difference between the measured and actual speed is only about 1.5 %.

The absolute value of the frequency shift is dependent on the frequency used. For this reason, it is preferable to use higher operating frequencies for recording low speeds and lower operating frequencies when measuring high blood velocities. **In practice, frequencies from 1 MHz to 10 MHz are used.** At blood flow rates from 1 cm/s to 500 cm/s, there is a frequency shift in the range of audible sound which allows sound in addition to the video recording.

1.1.4.1 Rheology – patterns of blood flow through blood vessels

Blood, a non-Newtonian fluid, has a laminar flow under physiological conditions. We can imagine the flow in a single direction as sliding concentric cylinders parallel to the vascular endothelium where the velocity of blood components in each cylinder is the same. The slowest moving is the layer of blood adjacent to the vascular endothelium. The closer to the center of the cylinder, the faster the layers move. The velocity of the other layers of blood gradually increases to a maximum in the blood vessel axis. The

laminar flow results in a cylindrical or parabolic profile where the rates of individual blood cells differ only slightly.

The frequency shifts according the the basic Doppler effect (see above). The smallest frequency shift is created by the slow moving cells at the edge of the vascular lumen. In contrast, the largest frequency shift is reflected by the US waves from fast moving blood cells in the center of the blood vessel. **Graphic recording of laminar blood flow is then a narrow frequency spectral curve with a small spectrum of fast moving blood flowing through the blood vessel**. A narrow frequency line spectral curve with a characteristic shape appears on the monitor.

In the case of **turbulent flow**, the erythrocyte velocity spectrum is considerably broader and the linear narrow frequency spectral curve expands on the monitor until it completely disappears. The area under the curve is filled with a number of spectral velocities.

1.1.4.2 Continuous Doppler

In vascular diagnostics, the first clinically used Doppler method was **continuous Doppler (Continuous Wave Doppler – CWD**). Continuous Doppler uses an unmodulated wave of 4 MHz or 8 MHz US signal that is continuously transmitted and received.

The probe for continuous Doppler scanning contains two crystals (piezoelectric transducers). One is a permanent transmitter and the other continuously receives reflected signals. The crystals are positioned in the probe so that the transmitted and received beams overlap in a sensitive area a few centimeters long. The US signal is transmitted and scanned continuously. It is more powerful than the pulse Doppler (see below), but as the sum of all the tissues through which the signal passes, it lacks precise spatial focusing of target structures. The US waves are converted into electric current by the piezoelectric transducer and an audible sound is heard as a stereoacoustic signal from a speaker or headphones. Computerised processing of the electrical signal leads to a graphic record of blood velocity simultaneously appearing on the monitor. This is a Doppler record of the velocity of blood through the vessel.

Inexpensive CD systems, without 2D image, are equipped with small pencil probes (4 MHz and 8 MHz). Pocket Dopplers serve as an orientative functional examination of the veins in the standing or lying patient. **Pencil transducers**, also called CW Doppler probes, are utilised to measure blood flow and speed of sound in blood. This probe has a small footprint and uses low frequency (typically 2–8 MHz).

Their great disadvantage however, is lack of spatial resolution. This prevents blood velocity measurement at a specific 2D image location. All the vessels located in the longitudinal axis of the US beam at different depths and sites are scanned at the same time and the resulting acoustic signal is mixed.

Since continuous Doppler has no limit for frequency shift, **its main function is accurate measurement of high blood velocities** in cardiology and detection of flow in superficial blood vessels.

1.1.4.3 Pulsed Doppler

Current US devices for vascular investigation use Pulsed Doppler (**Pulsed Wave Doppler, PWD)**, where one piezoelectric element in the probe alternately transmits

and receives the US waves to and from the tissue. These are reflected from the interface of different densities (blood/tissue, tissue/tissue) and received by the same transducer after a short delay. At a constant rate of US propagation in soft tissues, the time between transmission and pulse reception is directly proportional to the vessel's distance from the probe. The Pulsed Doppler system, therefore, allows precise determination of the depth the reflected signals are coming from. For this reason, Pulsed Doppler can be used to select a site in a particular vessel and place there a sampling volume (**measurement volume, sampling volume**) from which the Doppler signal is recorded. This is the main difference to continuous Doppler.

The low signal-to-noise ratio of Pulsed Doppler however, precludes its use for very slow flow rates. Another ultrasound mode should be used here – Power Doppler (see Chapter 1.1.4.6).

1.1.4.4 Color Flow Mapping (CFM, color Doppler)

In color flow mapping, the blood flow record of pulse Doppler is superimposed on a 2D image (B-mode) in real time. As a result, CFM has the limitations of both modes B-mode and PW Doppler (see above). The CFM allows a color display of the hemodynamics, i.e., different velocities and direction of blood flow in real time.

Pulse Doppler detects the velocity and direction of blood flow in several sample volumes simultaneously. The number of sample volumes is determined by the **size of the CFM display box** which constitutes a part of the 2D image in gray scale (B-mode). The size of the display box is determined by the machine settings. Selecting a too large box for CFM, especially in terms of width, is associated with reduced spatial resolution.

To identify vessels through CFM and PW Doppler, it is important to set the correct **pulse repetition frequency (PRF)** emitted by the probe. This must match the velocity of blood flow in the vein examined. For a slow velocity, set a lower PRF and a higher PRF for a fast flow rate. If the PRF is set too low, a **frequency error (ambiguity)** is generated. Frequencies (blood flow rates) that exceed the set frequency and thus speed range of the machine cause the computer to automatically re-set to the zero line, i.e. to negative values. These frequencies (speeds) are then relabelled (aliasing) as the opposite (negative or positive) frequency, indicating the opposite direction of flow. If the PRF is set too high, there is a **spatial error**. At the location of a slow-flowing vessel (the flow is slower than the PRF setting), no blood flow appears. There is an error in the identification of the vessel, i.e. the vessel can be overlooked as another anechoic structure (**Figure 1.2**).

The resulting CFM is a color flow map. This, as mentioned, is superimposed on a 2D image (B-mode), which forms the image background at the site of the blood flow investigation, and is defined by the CFM box boundaries (**Figure 1.3**). Blood flow is represented by points (voxels, pixels) in blue and red with intensity of color showing blood velocity – red showing blood flow back to the probe and blue flowing away from the probe. The quality of the 2D image covered by a color map is however somewhat poorer in places than without a map.

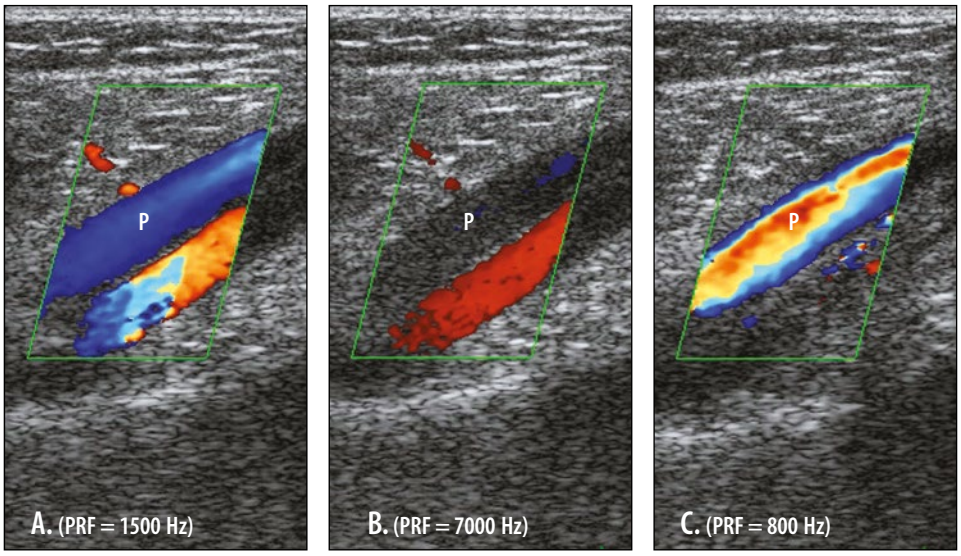

Fig. 1.2 *Pulse Repetition Frequency (PRF) on a US device for the investigation of the popliteal vein (P) – **A.** proper PRF setting (1500 Hz, corresponding to a flow velocity of ± 15 cm/s) which corresponds to the measured blood flow velocity in the vein, is a monochromatic, homogeneous flow pattern, **B.** PRF setting too high (7000 Hz, corresponding to a blood velocity ± 68 cm/s), at the site where the vessel lies, no blood flow is shown (CFM spatial error), **C.** too low PRF setting (800 Hz, corresponding to a flow rate of ± 8 cm/s), higher frequencies corresponding to faster blood flow in the blood vessel axis are displayed in the opposite (red-yellow) color coding (frequency error, CFM aliasing) an apparent change in direction of blood flow in high-velocity areas, producing flow that appears to be backward*

1.1.4.5 Duplex and triplex ultrasound examination

By combining 2D imaging in real time with pulse Doppler, we get a duplex ultrasound scan (**combination of B-mode and pulse Doppler**) (**Figure 1.4**). The gray scale 2D gives the morphological image but no movement. The color Doppler provides an image of blood velocity. The two displays overlap on the screen, so-called duplex.

The blood flow measurement point can be precisely selected by adjusting the sample volume in B-mode. Duplex sonography allows the B-image to determine the location and size of the sample volume, from which spectral readings are shown for blood velocity in the scanned vessel. The tissue architecture and recorded blood flow curve in real time are displayed for the selected site

The term **color duplex sonography** is used to indicate a 2D gray scale image which is integrated with color flow mapping (**B-mode and CFM**). It is also referred to as **Doppler color sonography**.

By combining the graphic signal of pulse Doppler with B-mode and CFM, we arrive at the **triplex US investigation (B-mode + pulse Doppler + CFM)** (**Figure 1.4**).

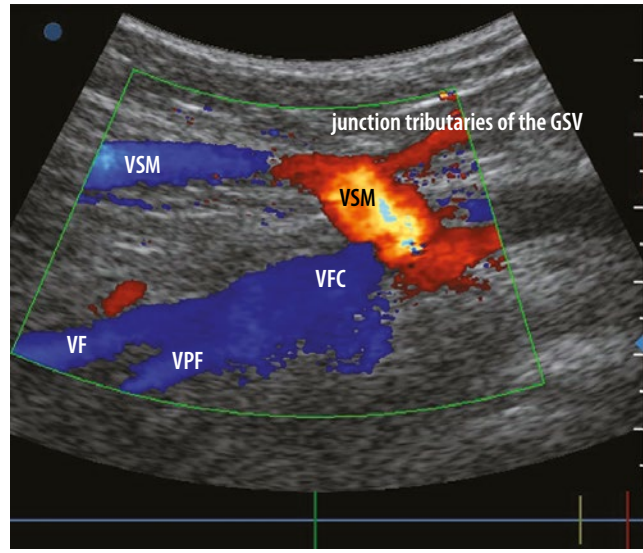

Fig. 1.3 *Color map of the SFJ on a background of a 2D US image – color flow map (CFM): in the box for the CFM display (green) the velocity and direction of the blood flow are displayed using PWD, blue indicates blood flow away from the probe, red, blood flow towards it, blood flow rate increase reflected as lighter shades of both colors. VSM – vena saphena magna, VF – vena femoralis, VFC – vena femoralis communis, VPF – vena profunda femoris*

Linear or sector probes are used for examining the area of the iliac veins, groin, thigh and popliteal vein using 4–5 MHz frequencies and for lower leg and ankle 5–10 MHz (Elias, 1998).

1.1.4.6 Power Doppler

Other terms used are: **Color Doppler Energy, Ultrasund Angiography, Color Amplitude Imaging (CAI), Color Power Angiography (CPA)**. The basis of this technique is reflected impulses with a frequency shift but the color shows only the integrated power of the reflected signal, with the frequency shift ignored. Simply put, the intensity of the color corresponds to the number of moving cells, regardless of direction and speed of movement. Since Power Doppler is not as dependent on the angle of incidence (the angle that determines the direction of blood flow and incident US waves) as other Doppler methods, it can be used to investigate less perfused tissues and organs, especially in slow-flowing sites.

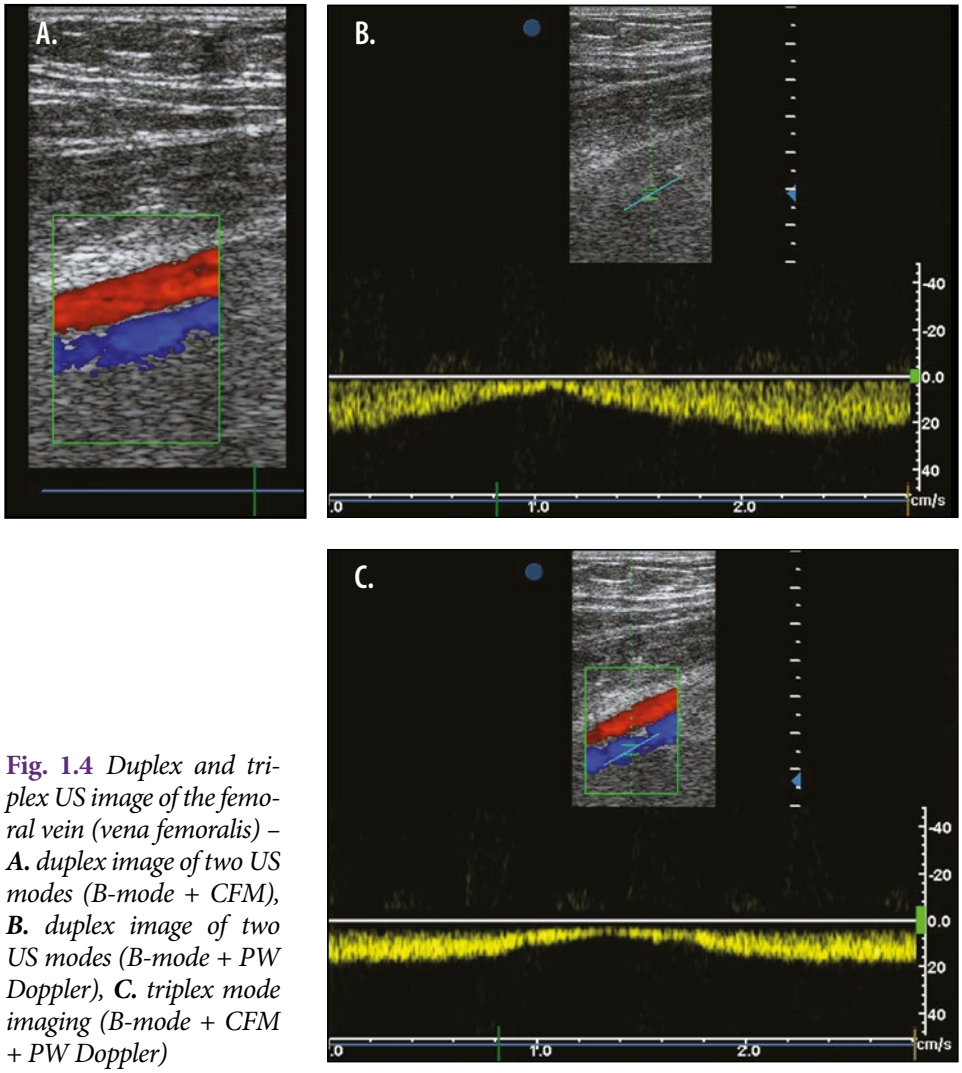

Fig. 1.4 *Duplex and triplex US image of the femoral vein (vena femoralis) –* ***A.*** *duplex image of two US modes (B-mode + CFM),* ***B.*** *duplex image of two US modes (B-mode + PW Doppler),* ***C.*** *triplex mode imaging (B-mode + CFM + PW Doppler)*

1.2 Clinical use of ultrasound in phlebology

Practical ultrasound examination is simple at first glance but it comes with a few challenges. You need to familiarise yourself with the device, find which knobs are most important and test the various ways for comparing the image of the same environment/site/medium while changing a single image parameter. Once you have found the appropriate setting for a specific parameter, you need to save it and investigate it with changes to the next. This may initially be laborious but it will allow you to become accustomed to the instrument and gain more from its imaging capabilities. Manufacturer specifications do not have to comply with specific clinical practice and user manuals are often poorly designed. It stands to reason also hat you need to investiga-