Ultrasound-Guided Peripheral Nerve Blocks

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Foreword

It is my great pleasure and privilege to introduce another volume on peripheral nervous system ultrasound anatomy, peripheral nerve pathology imaging, and ultrasound-aided regional anesthesia and pain management edited by Enzo Silvestri, Fabio Martino, and Filomena Puntillo.

Ultrasound is an emergent imaging modality that is widely used to assess peripheral nerve neuropathies. Among its many features, it is the only imaging modality that is able to perform dynamic evaluations of the soft tissues related to the musculoskeletal system and without patient exposure to ionizing radiation. Also, in expert hands, ultrasound enables the precise guidance of needles within soft tissues and joints, for use in regional anesthesia for a wide range of nerve blocks and for interventional pain management for relief of acute, chronic non-cancer, and cancer pain.

The book consists of three parts. In the first one, general knowledge on ultrasonography and peripheral nervous system ultrasound anatomy is presented. In the second part, concepts of nerve pathology and nerve entrapment syndromes are discussed with an emphasis on the most appropriate use of each imaging modality. In the third part, ultrasound-guided nerve blocks are pictorially presented offering point-by-point checklists for each procedure together with detailed anatomic schemes.

I would also like to emphasize that this handbook is based both on data obtained from the literature and the daily experience of authors who are all recognized opinion leaders in radiology, anesthesiology, and pain medicine. It therefore describes different approaches for the same procedure, allowing the reader to select the most suitable for the particular application.

I would like to thank and to congratulate most sincerely the editors and the authors for their efforts, which have resulted in this comprehensive but well-balanced and very readable text, completed with a remarkable ultrasound-guided nerve blocks section and a large series of dedicated didactic schemes.

This book will be of great value to both anaesthesiologists and radiologists, with a different level of experience, ranging from the physician in training to the one who already performs the treated procedures with traditional technique and want to become familiar with US guidance. It will provide

them with the state-of-the-art knowledge in the specific fields of peripheral nerve sonoanatomy and ultrasound-aided regional anesthesia and pain management.

I am confident that it will meet the same success with the readers as the previous volumes published in this series.

Genova, Italy Giacomo Garlaschi

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Fundamentals

1

Enzo Silvestri, Silvia Perugin Bernardi,
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1.1 Basic Principles

Ultrasonography (US) is one of the most widely used imaging technologies in the first-level study of each human body structure, including soft tissue components of the musculoskeletal system and nerves. It is quick, portable and free of radiation risk, and, thanks to its high sensitivity and image resolution, its applications are continuously increasing.

Furthermore, US allows to acquire the images in 'real time', thus providing instantaneous visual guidance for many interventional procedures and reducing the risk of complications.

Rapid advances in transducer technology (broadband and high-definition probes), development of tissue harmonic imaging (THI) systems, new dedicated software and reconstruction algorithms (compound imaging, steering-based imaging, extended field-of-view imaging, three-dimensional imaging, sonoelastography), together with the possibility of a dynamic analysis of tendons, muscular structures and nerves, resulted in increased diagnostic performances and have opened new fields of applications in interventional procedures including those for regional anaesthesia and pain management.

US examination is relatively operator dependent, and it presumes a good knowledge of the physical principles on which it is based and the technical properties of the available equipment.

In this chapter, we describe some of the fundamental principles and physics underlying US technology.

1.1.1 Ultrasound Wave Properties

Ultrasonography is based on the use of acoustic waves with frequencies higher than the human hearing range (>20 kHz).

Sound waves can be described in terms of their **amplitude** (measured in decibel), **frequency** (measured in cycles per second or hertz), **wavelength** (measured in millimetre), **period** (the time interval in which each oscillatory phenomenon is reproduced), **velocity** (greater in rigid or less compressible materials, lower in air, water and soft tissues), **power** (measured in watts) and **intensity** (measured in watts per square centimetre) (Fig. 1.1).

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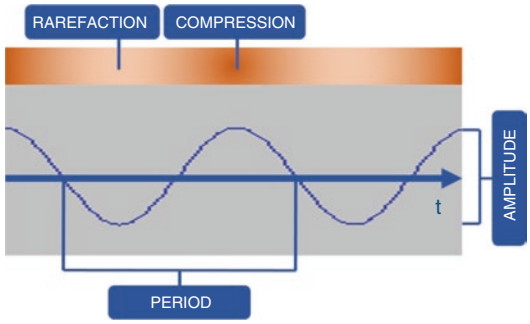

Fig. 1.1 The graphic demonstrates the oscillatory behaviour of the ultrasound waves propagating in tissues

1.1.2 Interactions Between US and Anatomical Structures

In ultrasonography, high-frequency sound waves are generated and transmitted through the body by a transducer.

Ultrasound transducers (or probes) contain multiple piezoelectric crystals which are electronically interconnected and vibrate in response to an applied electric current (piezoelectric effect). So, it produces not only the US but is also able to detect the returning echo produced by the interaction with tissue target, converting them back into electrical signals and then encoding into images.

Crossing anatomical structure, sound wave is subjected to varying phenomena that contribute to US image formation: attenuation, absorption, reflection, diffusion, refraction and divergence.

Attenuation (expressed in units of decibels per centimetre): it is reduction in the intensity of a signal through tissue. It depends on US frequency (MHz) and the distance travelled (cm): the higher the frequency and the distance, the greater the attenuation. So, the high-frequency US allows a better image resolution, but the beam is more attenuated. The implication, in the clinical practice, is that the high-frequency linear-array probes (7–13 MHz) are used for the study of superficial structures (tendons, muscles, ligaments) and the low frequency curved-array probes (2.5–5 MHz) for deep organ evaluation. The main causes of attenuation are absorption, reflection, refraction, diffusion and divergence.

Absorption: the acoustic energy is transformed into thermal energy of the tissue, according to US beam frequency and intrinsic characteristics of the medium (low in liquids, intermediate in soft tissues, high in bones and air).

Reflection and refraction: these are the deviation of the transmitted beam from the incident beam direction. The energy can be reflected back to the transducer (reflection) or deviated through the interface (refraction). These phenomena depend on the angle of incidence (the orientation of sound wave relative to the surface) and the different acoustic impedances between two materials.

The reflection increases when the US beam approximates to the angle of incidence, that is, equals to zero (when the US beam is perpendicular to the surface) while the refraction decreases and vice versa.

The acoustic impedance is an intrinsic property of tissues which expresses their resistance to the passage of the US beam (expressed in units of Rayls = kilogrammes per square metre per second).

So, we have the strongest detected returning echoes in clinical sonograms (*specular reflection*) when the angle of incidence equalled to zero and the difference in acoustic impedance between two materials is high.

Diffusion: according to the characteristics of the interface (homogeneous or irregular), the reflection can be *specular* (mirror-like) or *diffuse.* The latter one is the scattering of the reflected US beam in all directions because we have multiple interfaces (i.e. parenchymal organs).

Divergence: when the US beam is not focused, it diverges distally with consequent reduction of the depth resolution (Fig. 1.2).

1.1.3 Images Formation

A pulse-echo type of measurement allows to obtain ultrasonographic images.

The transducer is composed by a number of piezoelectric crystals assembled in a linear or curved disposition. Each crystal is excited by electrical pulses (*reverse piezoelectric effect*) and converts electricity (electrical energy) into sound (mechanical energy). The combination of these

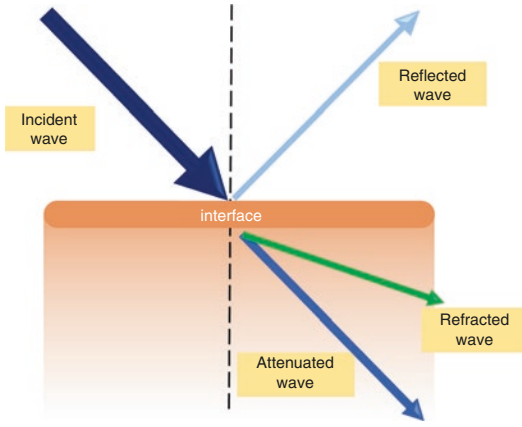

Fig. 1.2 Scheme of the physical phenomena contributing to US image formation

multiple beams generates the US beam. When the latter one is reflected to the transducer by tissue structures (returning echoes), it is converted back into electrical pulses (*piezoelectric effect*) and then in the image presented on the screen.

Currently, transducers contain a range of ultrasound frequencies (bandwidth) instead of a single fundamental frequency.

The generation of US images requires a complicated acquisition and display of ultrasound pulse-echo data. The broadband transducer generates a sequential series of focused beams all in the same plane (scan plane). Each set of target data from a single pulse transmission is placed in the image, as acquired along a line. All tissues in the scan plane are interrogated by these beams, and each real-time image frame is composed of a set of parallel or sector lines representing the positions of the interrogating beams in the patient. Computer algorithms are used to fill in between the image lines so that the image appears continuous.

When the transmitted US pulse encounters internal tissue targets, part of its energy is deflected (reflected or scattered) back to the transducer (the echo). Because pulse-echo imaging techniques employ the same transducer for both sending and receiving US pulses, only echoes travelling in the direction of the transducer have any chance of being detected.

The main pulse-echo parameters used in the formation of images include echo amplitude and target spatial position. Echo amplitude is encoded into shades of grey (greyscale imaging), with the lighter shades representing higher amplitude echoes.

The depth of the target along the direction of the beam is accurately calculated from a pulse time-of-flight measurement. Assuming US propagation velocity is fairly constant from tissue to tissue (1540 m/s), the time between beam transmission and echo reception is used to determine the exact internal spatial location of all tissue targets.

The image quality is represented by two important parameters: *spatial* and *temporal resolutions.*

The first one refers to the capability to distinguish two adjacent points, along (*axial resolution*) or perpendicular to the axis of the beam (*lateral resolution*).

Temporal resolution is linked to real-time identification of anatomical structures according to the pulse repetition frequency (PRF) and the frame (number of encoded images per time unit).

Visualization Systems:

- Amplitude mode (*A-mode*): it is the simplest form of display. It is a diagram in which echo amplitude is shown according to tissue depth (echo time of flight).
- Time-motion mode (*TM-mode*): echoes returning from moving structures are displayed depending on the time. It is used in cardiac US evaluation.
- Brightness mode (*B-mode*): it is a greyscale tomographic imaging (Fig. 1.3).

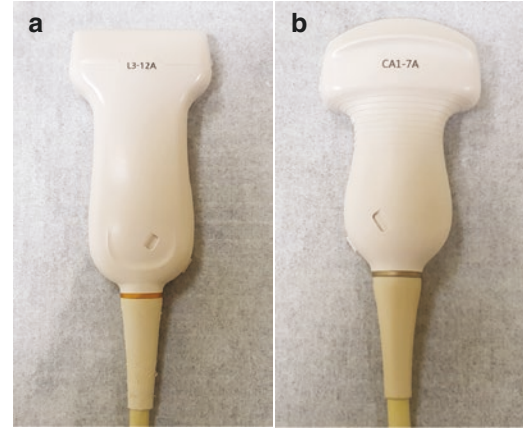

Fig. 1.3 **(a)** Linear-array probe (5–12 MHz). **(b)** Convex-array probe (2–5 MHz)

1.1.4 Artefacts

A detailed description of US artefacts lies outside the aim of our handbook. However, the errors in image display are important to know because it can be induced in an uncorrected interpretation of clinical findings.

The artefact can be linked to improper scanning technique or the physical characteristic of the US beam.

Anisotropy is an artefact that originates from a loss of echogenicity in structure. It is strictly related to US beam angle of incidence. If the US beam is not perpendicular to linear structures, the reflection is not specular, and so the returning echoes have low intensity: the structure wrongly appears more hypoechoic. The knowledge and the capability to correct the anisotropy artefacts are very important in order to achieve a high diagnostic accuracy and ultimately an optimal management of patients.

US Equipment: Hands On

If the tissue target is superficial and small (i.e. nerve), we have the necessity to increase the spatial resolution and image definition.

So, in order to improve image quality and diagnostic performances, some important parameters can be adjusted before and during US examination.

1. *Probe selection*: high-frequency linear-array probes, operating with frequencies of 10 MHz or more.
2. *US beam focusing*: it determines the number and pattern of focal zones. The zone in which the width and thickness of the US beam are reduced dynamically.
3. *Gain adjustment*: it optimizes echo intensity at different levels of depth.
4. *Zoom*: it better visualizes small and thin structures.
5. *Dynamic range adjustment*: it must be reduced in order to enhance the contrast resolution. In fact the dynamic range is inversely proportional to contrast resolution.

1.2 Doppler and Sonoelastographic Imaging

1.2.1 Doppler Imaging

Doppler effect is a change in frequency of the reflected ultrasound waves backscattered from the structure that is in motion, and it is based on an essential principle: the sound frequency of a target changes as the target travels towards or away from a point of reference. In particular, when the US beam, produced by the probe, is transmitted into a vessel, the frequency of the received wave is different from that of the transmitted wave because the source (red cells) moves relative to the given receiver (probe). Doppler ultrasound describes a frequency shift between an emitted ultrasound beam and the received echo.

The change of frequency detected between the transmitted and the received US frequencies is named 'Doppler shift'. The received US frequency would be higher if the direction is towards the receiver and lower if the direction is opposite. The equation which describes this phenomenon is:

$$\Delta f = (2\, fo\, v/c) \times \cos \alpha.$$

C: speed of ultrasound in soft tissues
V: speed of erythrocytes
fo: emitted frequency
$f1$: the frequency of the reflected ultrasound
$\Delta f = fo - f1$: Doppler shift
α: angle between the direction of the movement and the direction of the US beam

From the Doppler equation seen above, we see that the Doppler shift is influenced by the following factors:

1. The frequency of the ultrasound beam (*f*) used to interrogate flow
2. The angle of the ultrasound beam to flow direction (α)
3. The velocity of flowing red blood cell aggregates (*V*)

The Doppler shift of the moving red blood cells is continuously monitored to produce the Doppler signal; it is in the audible range and can thus be heard. The resulting sound is distinct and provides feedback to the operator.

In Doppler measurement, the Doppler angle is very important: the beam incidence should be from 0 to 89° because at 90° there is no signal (cos 90° = 0). Practically, 30–60° offers the best Doppler angle (Fig. 1.4).

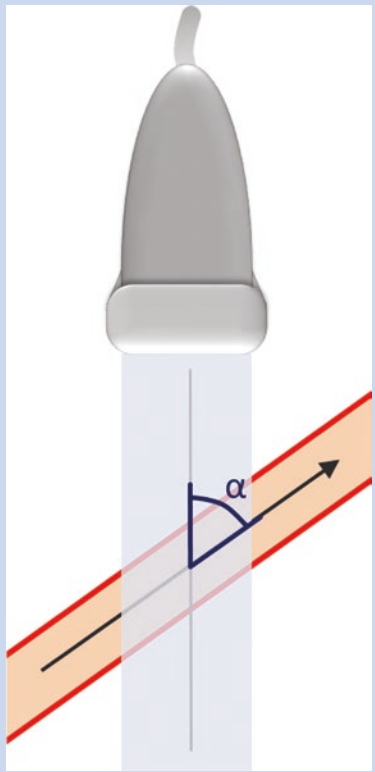

Fig. 1.4 Angle of Doppler insonation (α). Given a flow direction, ideally the transducer should be parallel to blood vessel but that is not possible. 30–60° offers the best Doppler angle

1.2.1.1 Signal Processing

Changes in frequency are expressed in a graphic mode with spectral representation which comprise the feedback signal frequency (longitudinal axis) versus time (transverse axis). The analysis of the frequency is performed using the fast Fourier transform, a physical phenomenon that allows variations in the amplitude of the wavelength obtained by the device to be displayed in the range of frequencies.

Some parameters that the radiologist must know and adjust influence the quality of the spectral analysis (Figs. 1.5, 1.6 and 1.7):

- *Pulse repetition frequency*: determines the number of pulses originated in the machine; they differ if we use B-mode or Doppler at the examination. High PRF detects high speed; low PRF detects slow flow, and if it detects high flows, aliasing occurs.
- *Size of the sample box*: the modification of the volume of the sample produces effects on the spectrum; if the size is too big, you get signal record below the spectrum. It marks the point at which the flow rate is determined. Sample volume should be 1/3 diameter of the artery. For very small blood vessels and veins, expanding sample volume is used. Too wide sample volume in the arteries causes spectral expansion.
- *Gain control*: if it is augmented, it can increase noise in the spectrum background and overestimate the velocity, and on the other hand, if it is reduced, the spectral record is not well demonstrated and can underestimate the velocity.
- *Angle control*: the angle of insonation is critical in estimating the correct velocity of the vessel of interest. When the angle is correctly adjusted under 60°, the spectrum is better delineated. When the insonation includes an angle greater than 60°, there is spectral broadening.
- *Wall filter use*: it annuls the signals on the wall or out of the blood vessels, ignores frequencies below a threshold and can be controlled by the operator. When a low-speed flow is explored by a very high filter, it could suppress the Doppler signal.

- *Flow parameters*: they provide diagnostic thresholds and are derived from the spectral frequency. The major representatives are resistive index (RI) and pulsatility index (PI) not influenced by the angle of insonation but useful to determine flow resistance in the vascular system. The operator can calculate them using the spectrum (Fig. 1.8).

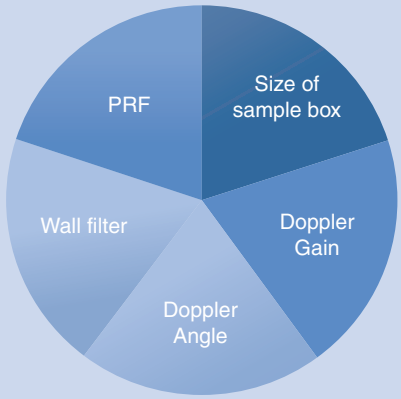

Fig. 1.5 Technical factors influencing the quality of the Doppler spectrum

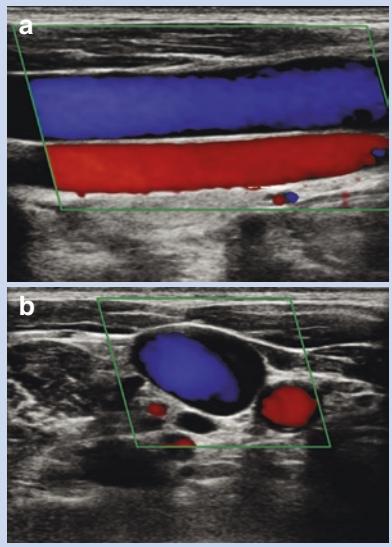

Fig. 1.6 Colour Doppler imaging shows the carotid artery (red) and jugular vein (blue) at the level of the neck in a longitudinal scan (**a**) and in a transverse scan (**b**). Especially in imaging soft tissues, identifying neurovascular bundles with US colour Doppler technique can be very useful to correctly localize tendinous and muscular structures

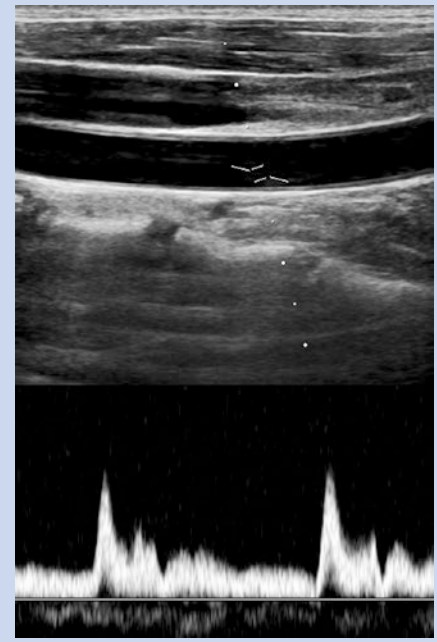

Fig. 1.7 Ultrasound spectral examination of the flow of the carotid artery

Resistive Index (RI)

$$RI = \frac{\text{max V} - \text{min V}}{\text{Max V}} = \frac{\text{S-D}}{\text{S}}$$

Pulsatility Index (PI)

$$PI = \frac{\text{max V} - \text{min V}}{\text{Mean V}} = \frac{\text{S-D}}{\text{Mean}}$$

Fig. 1.8 Resistive index and pulsatility index equation. *MaxV* maximum velocity, *minV* minimum velocity, *S*, sistole; *D* diastole

Doppler ultrasound is an imaging technique that combines anatomical information derived using US pulse-echo techniques with velocity information derived using Doppler techniques to generate colour-coded maps of tissue velocity superimposed on greyscale images of tissue anatomy.

The most common use of the technique is to do the accurate noninvasive evaluation of the blood flow movement through the heart, arteries and veins, but it may also be used to image the motion of solid tissues such as the heart walls. As the name implies, US Doppler technique uses the Doppler effect to assess how blood flows through the major blood vessels.

There are three methods of analysis:

- *Qualitative analysis*: evaluating the presence, site and direction of the blood flow
- *Quantitative analysis*: evaluating flow velocity and flow rate
- *Semi-quantitative analysis*: evaluating the spectrum of the wave frequencies

There are two types of technical mode of Doppler imaging in medicine:

- *Continuous-wave Doppler* (CW Doppler): it uses two separate crystals, one transmitting and the other one receiving the reflected waves. The first crystal transmits a continuous signal at a known frequency, and the other crystal receives the returning echoes and records their frequency. Consequently, the US machine cannot determine which sound pulse was frequency shifted and, therefore, cannot precisely define the location of the moving target. In conclusion, this technique is employed in detection of blood flow but does not give information about direction, depth and velocity of flow.
- *Pulsed wave Doppler* (PW Doppler): one transducer is used as a transmitter and as a receiver. The probe produces US beams in pulses, alternating the transmission and reception. This has provided the means of detecting the depth at which a returning signal has originated. The depth can be positioned at any point along the axis of US beam referred to as the 'sample volume'. The position of the sample volume is decided by the operator. When the US beam is transmitted into tissues, it travels for a given time until it is reflected by a moving red cell; then, it returns to the probe over the same time interval but at a shifted frequency. Calculating the total transit time, the US machine is able to measure the distance of the sample volume. In respect to the CW Doppler, PW Doppler is able to evaluate the depth from which the returning echoes originate, but it cannot correctly depict higher velocities (blood flow velocity measurements are limited to the physiologic range, usually around 1.5 m/s).
  PW is based on three technical parameters: *high pulse repetition frequency* (the difference between successive burst of incidence ultrasound beam), the optimum *transducer frequency* (low frequency for deeper structures and high frequency for superficial structures) and the correct *insonation angle* (<60°).

PW Doppler is usually combined with a 2D, real-time, B-mode scanner to form what is known as a duplex scanner.

PRF scale (repeated pulsing frequency) is the number of pulses per unit of time that is transmitted to the blood vessel, and when it is too long relative to the velocity of the blood flow and it will not be possible to determine the direction of blood flow, we have the aliasing phenomena. In particular, aliasing occurs when the velocity is more than one half of the PRF; in this case, velocities above this limit will be displayed on the tracing opposite to the true direction of blood flow. To correct for aliasing, the operator can increase the PRF or increase the angle between the US beam and the flow direction towards perpendicularity.

Colour Doppler: It is an ultrasound system in which the echo signals received along a series of locations in an ultrasound beam width by applying transmit-receive pulse signals are called pulse packets. The energy of the returning echoes is displayed as an assigned colour; by convention, red for echoes that flow towards the probe and blue for echoes that flow in the opposite direction, away from the transducer. US machine displays coloured blood flow superimposed on a greyscale image, thus allowing simultaneous visualization of anatomy and flow dynamics. Brighter shades in colour conventionally depict faster flow.

To optimize the colour Doppler evaluation, it is crucial to set the US beam at an optimal angulation <60° in respect to the vessel, basing on the physical Doppler equation (cos 90° = 0).

Power Doppler: It is a type of colour Doppler, more sensitive to blood flow compared with conventional colour Doppler, ignores the velocity and the direction of flow and simply estimates the strength/amplitude of the Doppler signal detected from each location. Power Doppler shows small vessels and slow flow rates; indeed it is most commonly used to evaluate low-velocity microvascular flow in soft tissue imaging. Power Doppler is extremely sensitive to the movement of the probe, which produces a flash artefact.

So the advantages of power Doppler versus colour Doppler imaging are:

- More sensitive to flow states.
- Angle effects are ignored.
- Aliasing artefact is not applicable to power Doppler imaging.

The disadvantages are:

- Values of velocity and the direction of blood flow cannot be assessed.
- Flash artefact: because of more averaging of information at slower frame rates, slow-moving soft tissue signals appear as flash artefact.

Power Doppler should be optimized while the probe is not in contact with patient's skin. The gain should be set at maximum level and then decreased up to the disappearance of all artefacts. Further, it is important to set low wall filters (WF) and pulse repetition frequency (PRF) between 700 and 1000 Hz in order to better evaluate low-velocity blood flows.

Combining with greyscale ultrasound, colour Doppler imaging and power Doppler imaging allow unique real-time evaluation of the regional blood flow, enabling a wide range of applications for the evaluation of soft tissues.

Often blood vessels are used in musculoskeletal imaging as anatomical landmarks. Colour Doppler and power Doppler are very helpful in detecting inflammatory diseases and neovascularity, possibly related to malignancy, and so they represent a useful tool also for the quick assessment of vascular anomalies and post-traumatic vascular lesions. It is also important to use colour Doppler imaging during a biopsy to ensure that major vessels are avoided.

1.2.2 Sonoelastography (SEL)

Sonoelastography (SEL) is a recently developed imaging technique which allows for qualitative visual or quantitative measurements of the mechanical properties of tissues. It is based on the principle for which, applying an extrinsic (mechanical or physical) stress, it is possible to induce changes in a determined tissue, depending on the elastic properties of the tissue itself; hence, qualitative and/or quantitative measurements of the elastic changes induced through the tissue could be obtained, usually by mean of an ultrasound transducer in clinical practice. The recent diffusion of SEL into commercially available ultrasound systems has promoted the development of many studies regarding the potential clinical applications of this technique in different clinical fields and, in particular, in the musculoskeletal system.

1.2.2.1 Elasticity: Basics Principles

The elasticity of a material represents its tendency to resume its original shape and size after being subjected to a deforming force or stress. Fluids resist a change in volume but not in shape: they have only 'volume elasticity'. Solids instead resist changes in volume and shape: they present

rigidity or 'shear elasticity', as well as volume elasticity. Viscoelastic fluids also exhibit elasticity in certain conditions.

It is essential that the terms stress and strain be defined because the elasticity of a material is described in terms of a stress-strain relation: the 'strain' is the relative deformation in volume or shape, produced by a force per unit area (called 'stress').

For a homogeneous isotropic solid, the ratio of stress-strain is a constant, called the 'modulus of elasticity'. A modulus measures the amount of force per unit area (stress) needed to achieve a given amount of deformation and usually is expressed in units of Pa. A higher modulus typically indicates that the material is harder to deform.

Three moduli are commonly used to define elasticity:

- *Young's modulus* (E) represents longitudinal elasticity and is defined by the ratio between the stress and the strain.
  Young modulus $E = S/e$.
- *Shear or torsion modulus* (G) represents transverse elasticity.
- Bulk or volume modulus (K) represents volume elasticity.

A stress determines two types of mechanical waves in the tissue:

1. *Compression wave* that compresses tissue little by little, inducing a displacement parallel to the propagation direction
2. *Shear wave* that is responsible of a slip of different tissue layers, relative to each other, inducing a displacement perpendicular to the wave propagation direction

The ultrasound elastography quantitative techniques do not directly measure the Young's modulus but the speed V of shear wave propagation.

The velocity V of the shear wave is related to shear modulus μ (shear):

$$\mu = r V^2 \text{ with } r = \text{tissue density}$$

The shear modulus μ is itself connected to the elastic modulus:

$$E = 3\mu$$

The measurement of the shear wave propagation velocity V (in m/s) allows to assess the elastic modulus E according to the formula:

$$E = 3\rho V^2$$

For computations, the tissue density is assumed to be constant and equal to 1000 kg/mm^3.

The shear modulus describes the response to shear forces, Young's modulus describes the response to linear stress (tensile stress) and bulk modulus represents the response (in all directions) to uniform compression; it is usual for values of shear and Young's modulus to be reported in the studies regarding the investigation of elastic properties of tissues by means of ultrasound.

So the aim of elastography is to assess tissue stiffness based on three steps:

1. Excitation: transmission of stress in a tissue (mechanical, vibrational, shear)
2. Acquisition: recording the signal induced by the tissue deformation due to the stress (RF or B-mode data)
3. Analysis/post-treatment: analysis of tissue strain induced by the propagation of the stress

Human body has a mechanical behaviour similar to a soft homogeneous and isotropic linear elastic material.

1.2.2.2 Modalities

There are several elastographic techniques depending on the difference in the stress application and the method used to detect tissue displacement and build the image (Figs. 1.9 and 1.10).

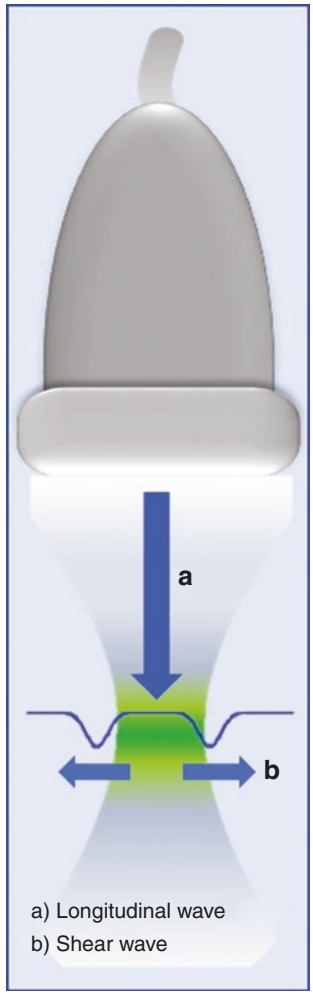

Fig. 1.9 Two different types of mechanical wave. (**a**) longitudinal wave; (**b**), shear wave

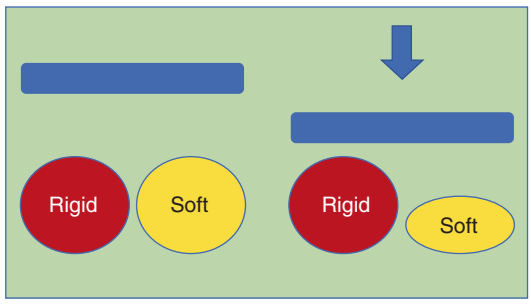

Fig. 1.10 The elasticity of rigid and soft materials

Two main types of SEL have become established in clinical practice, in particular for soft tissue evaluation:

- *Strain elastography*: it is also described as 'quasi-static elastography', 'compression elastography' and 'real-time elastography'. The stress is applied by repeated manual compression of the transducer, and the amount of tissue deformation (strain) relative to the surrounding normal tissue is measured, usually with a tracking algorithm working on the radio frequency data. The resulting data can then be used to form an image that is coded in colour or greyscale to show the pattern of strain, which is inversely related to tissue stiffness and can be assessed subjectively. These are qualitative data; however, regions of interest (ROIs) can be positioned over target areas in the screen in order to obtain semi-quantitative analysis (Fig. 1.11).
- *Shear wave elastography*: it is a very potential technique for the noninvasive quantification of tissue stiffness. Shear waves in the body can be induced by various methods, including physiological motion, external mechanical excitation or acoustic radiation force (by a focused ultrasound beam). Shear waves are transverse, they are rapidly attenuated by tissue, they travel much more slowly (between 1 and 10 m/s) and they are not supported by liquids of low viscosity. Using a real-time imaging modality such as ultrasound (but also magnetic resonance), the underlying tissue stiffness can be estimated measuring the produced shear wave speeds. Their speed is commonly expressed in metres per second (m/s); it is closely related to the modulus of elasticity of the tissue, and there is a simplified formula for converting between the shear wave speed and the elastic modulus of the tissue to locally quantify its stiffness in kilopascals (kPa). In contrast to strain elastography, this technique allows for the performance of quantitative analysis of the tissue stiffness. There are some

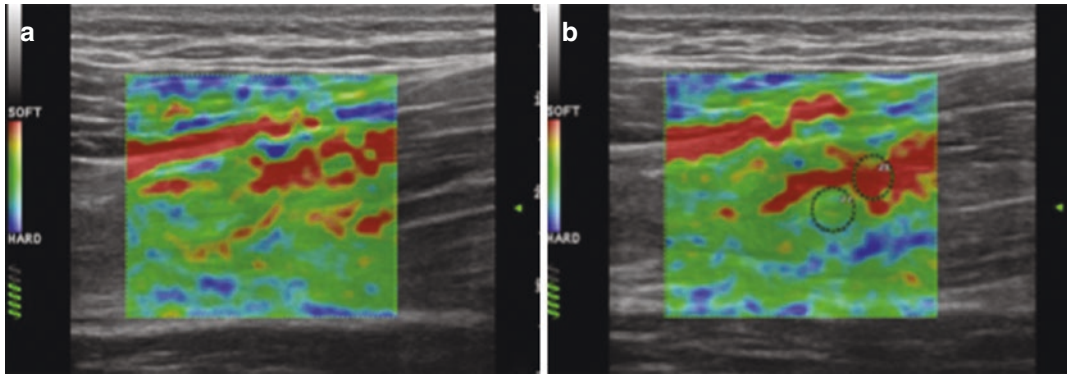

Fig. 1.11 Strain elastography. (**a**) Qualitative analysis: the modulus of elasticity of the soft tissue scanned in the B-mode image is represented by a superimposed colour-coded map in which (in this case) the lower values are depicted in red and the higher ones in blue; (**b**) it shows the possibility to perform also a semi-quantitative analysis of the strain elastogram with placement of two ROIs in order to take definite measurements of the Young's modulus of elasticity of the targeted tissue. The green-coloured spring-shaped figure shown in the left bottom of both the elastograms indicates that the pressure the operator performed with the transducer was appropriate to produce an adequate stress to get the elastogram

variations of this method in clinical practice, depending on the difference in the modality of stress application:

- *Transient elastography* (TE): it is a system developed and commonly used for liver fibrosis assessment, in which a mechanical piston within an ultrasound transducer is used to apply a push to the skin over an intercostal space. The speed of the produced shear waves into the liver, along the direction of the ultrasound beam, is measured in a way similar to M-mode.
- *Acoustic radiation force imaging (ARFI)*: in this technique, a focused ultrasound 'pushing' beam (with intensity below the threshold for bio-effects) is used to induce tiny displacements in soft tissue along its direction and generate orthogonal shear waves that propagate sideways in tissue. The shear wave speed or amplitude is detected by conventional ultrasound using tracking algorithms and is used to quantify the underlying tissue stiffness. Shear wave speed measurement could be made by a single small measurement box positioned by the operator within the tissue adjacent to the pushing beam (Fig. 1.12) and/or could be extended to sequential multiple pushing and measurement points in order to construct a colour-coded map of the shear wave speed, which is also quantitative with positionable ROIs (Fig. 1.13). ARFI images represent the spatial distribution of tissue stiffness.
- *Supersonic shear imaging (SSI):* it is a similar system which uses multiple acoustic radiation force impulses focused at different depths to create an extended cylindrical wavefront. These excitations are applied supersonically so that the shear waves generated from different depths constructively interfere adding each other's and dedicated ultrasound transducers could detect and measure them.

An overview of the different elastographic techniques is shown in Fig. 1.14, Tables 1.1 and 1.2.

1.2.2.3 Technical Considerations

In strain elastography, data acquisition and interpretation of elasticity images are largely dependent on the operators' experience and skills. SEL soft-

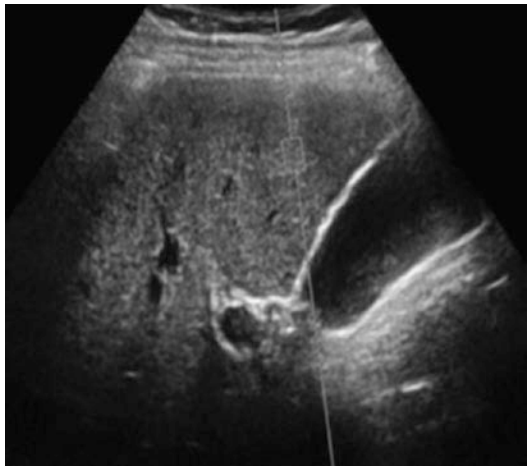

Fig. 1.12 Acoustic radiation force imaging (ARFI): shear wave speed quantification is obtained by a single small measurement box positioned by the operator within the tissue (fifth segment of the liver in this case) along the direction of the pushing beam. Data regarding liver segment, depth of the box placement and shear wave speed expressed in meters per second are represented right to the B-mode image

ware derive elastograms which usually depend on the changing probe pressure experienced during freehand scanning and on the individual capability of images' interpretation: consequently possible significant intra- and interobserver variability has to be taken in consideration. Further, this technique provides only qualitative and/or semi-quantitative analysis with elasticity data resulting from the relative stiffness of the targeted region and the remaining tissue area. Hence, this technical feature may significantly influence the clinical use of strain elastography in terms of reproducibility and accuracy.

In contrast, the quantitative nature of shear wave elastography is an advantage and seems to let this technique be more reproducible; the fact that the system displaces the tissue could improve consistency since the examiner does not need to move the transducer. The localized nature of the applied force should also improve the relationship between displacement and elasticity com-

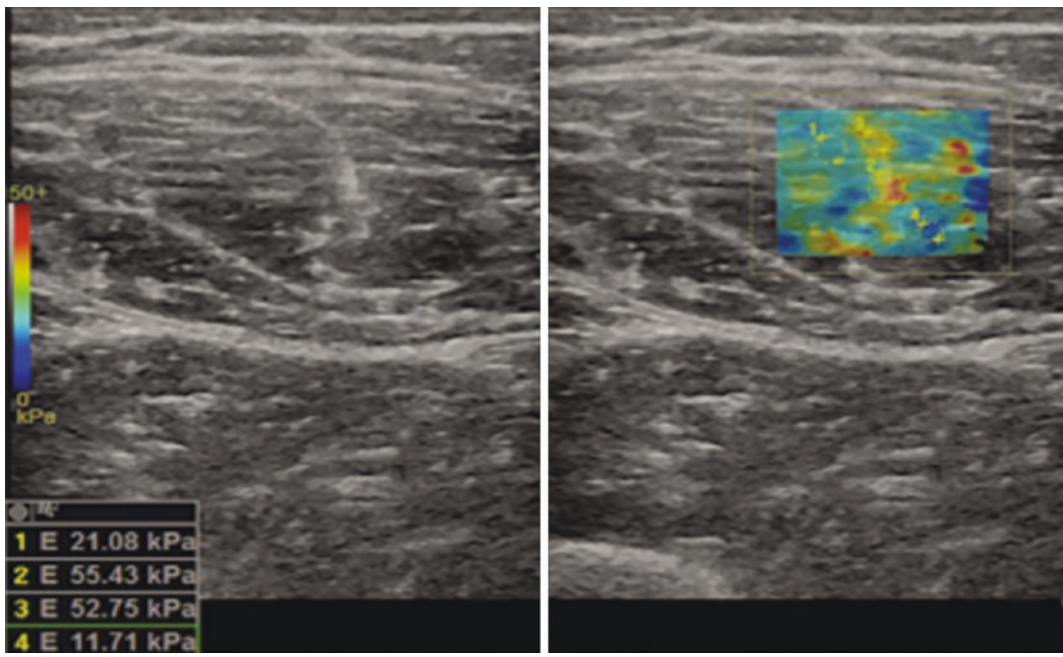

Fig. 1.13 Shear wave elastography of rectus femoris muscle: after the generation of the 'pushing' beam by the transducer, the values of the shear modulus in the targeted area are represented by mean of a colour-coded map set as represented by the coloured bar on the left of the screen. It is possible to get also a quantitative analysis of the investigated tissue by placing some ROIs (with modifiable dimensions) over the map and get the corresponding value at the left bottom angle of the screen. Note that, on the right elastographic map, the stiffer areas in the centre of the map correspond to the central rectus femoris aponeurosis

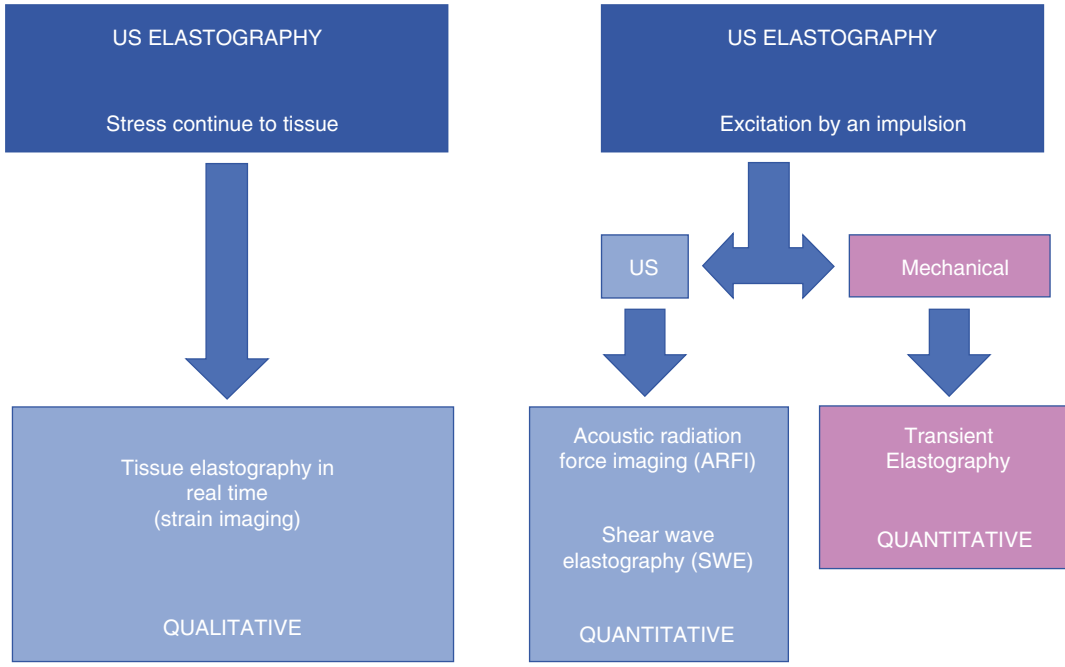

Fig. 1.14 Several elastographic techniques depending on the difference in the stress application and the method used to detect tissue displacement and build the image

Table 1.1 The technical characteristics of the different elastography modes

Elastography	Quasi-static		Dynamic	
Method	Strain imaging	ARFI	TE transient elastography	Ultra-fast shear wave elastography
Excitation mode	Mechanical (external compression) or physiological	Ultrasonic (radiation force)	Mechanical (pulse by an external vibrator)	Ultrasonic (radiation force)
Stress application	Surface or internal structure	Different depths	Surface	Different depths
Involved modulus	Young	Shear	Shear	Shear
Measured parameters	Displacement → strain	Shear wave velocity	Shear wave velocity	Shear wave velocity
Visualization	Temporal strain map	One fixed image	One measure (no image)	Temporal images (several/s)
Quantification	No	Yes	Yes	Yes

pared with applying the force at the surface, as well as improve contrast and spatial resolution. Despite the overall promising features of shear wave sonoelastography, in particular if compared with those of the strain elastography, some limitations have to be mentioned. Shear wave speed measurements using radiation force produced by a focused ultrasound beam can be dependent on transducer geometry, focusing depth, lateral tracking range and frequency of the shear wave used for imaging. Further, the shear wave speed in tissue is dependent on the shear modulus and its density, usually calculated by making some conventional assumptions which not always reflect the actual characteristics of the investigated tissue.

Sonoelastography is a very promising tool in addition to B-mode sonography and colour/

Table 1.2 Advantages and drawbacks of quasi-static and dynamic elastography

Elastography	Method	Advantages	Drawbacks
Quasi-static	*Strain imaging*	Easy	Operator dependent Applicable for superficial organ Qualitative method
Dynamic	*TE (transient elastography)* *Arfi Swe* *SWE (share wave elastography)*	Quantitative method Easy Validated for liver fibrosis Quantitative method Short breath hold Quantitative method Elasticity map	Limits for overweight and ascites Depth limited to 8 cm Longer breath hold

power Doppler techniques to evaluate stiffness changes in various soft tissue structures. Strain elastography provides quick, easier and qualitative or semi-quantitative measurements of such structures; shear wave elastography adds a more precise quantitative characterization with a much more difficult learning curve and a longer examination time.

1.3 Normal Anatomy

1.3.1 Peripheral Nerves

Peripheral nerves are usually made of nervous fibres (containing axons, myelin sheaths and Schwann cells) grouped in fascicles and loose connective tissue (containing elastic fibres and vessels). Each fascicle is encased by a proper connective sheath called *perineurium*. Inside the fascicle are a group of axons bathed in *endoneurial fluid*. Each axon has an insulating lining of *myelin*—a fatty material inside the *Schwann cells*. Between the fascicles and the outer nerve sheath, there is a fatty material called the *interfascicular epineurium* which houses the nerve vascular structure. The nerve is then wrapped in the main *outer epineurium*—an external sheath.

Clinical experience with ultrasound and improvements in technology have been helpful in the evaluation of peripheral nerve, and the improvements in Doppler sensitivity and power Doppler have made it possible to assess vascular changes within major nerve segments.

Peripheral nerve ultrasound, when compared to electrodiagnostic testing, adds the possibility to provide anatomic detail of the affected site without any discomfort. In fact ultrasound is a low-cost, quick and noninvasive imaging method, providing an excellent view of peripheral nerve anatomy as well as of surrounding structures. US provides high spatial resolution and the ability to explore long segments of nerve trunks in a single study, also allowing nerves examination in both static and dynamic conditions, during passive or active movements of the extremities.

US enables the identification of post-traumatic changes of nerves, neuropathies secondary to compression syndromes and inflammatory or neoplastic nerve lesions as well as the evaluation of postoperative complications, and it is increasingly used in anaesthesiology for regional anaesthesia.

Nerves present cable-like structures and have a distinct architecture consisting of fascicles and surrounding epineurium (Fig. 1.15).

In the transverse plane, the echo pattern is described as a ‘honeycomb’ aspect because tiny round and hypoechogenic areas representing the nerve bundles with hyperechogenic rims of the epineurium are visible.

In the longitudinal plane, nerves present as long, slim structures with a mixture of parallel

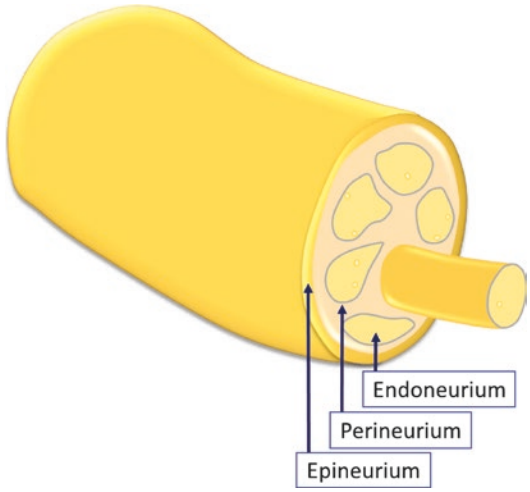

Fig. 1.15 Scheme of peripheral nerve illustrating its inner structure

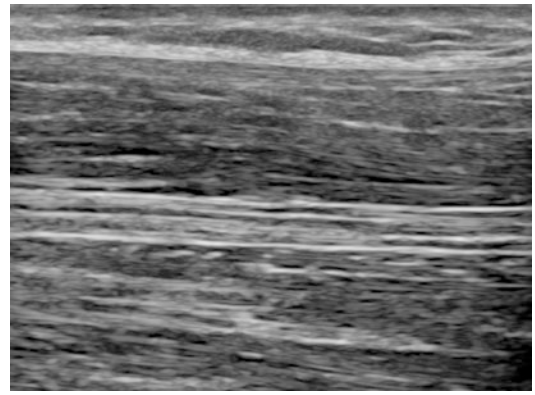

Fig. 1.16 Peripheral nerves. Longitudinal 3–16 MHz US image obtained over the median nerve (white arrows) at the middle third of the forearm. The nerve is made of parallel linear hypoechoic areas, the fascicles, separated by hyperechoic bands, the interfascicular epineurium

hyperechogenic lines, representing the perineurium, between two more prominent and also hyperechogenic layers of the epineurium. This image resembles that of an electric cable (Figs. 1.16 and 1.17).

The transverse image is much more frequently used in clinical practice, as it allows for the nerve to be examined by the so-called *elevator technique* which consists of finding the set nerve at a characteristic anatomic point and 'tracking it' either proximally or distally. In this way, it is possible to assess the nerve's shape, echogenicity and thickness and its relation to the surrounding tissues, the surface area of the nerve and its vasculature. If an abnormality is seen in the transverse view, the nerve should be examined in the longitudinal view.

The US aspect of nerves changes from hypo- to hyperechogenic as they are followed more peripherally for an increasing amount of connective tissue between the nerve bundles. It has been assumed that nerves are not anisotropic even if the property of anisotropy is seen in cases of nerves with large cross sections.

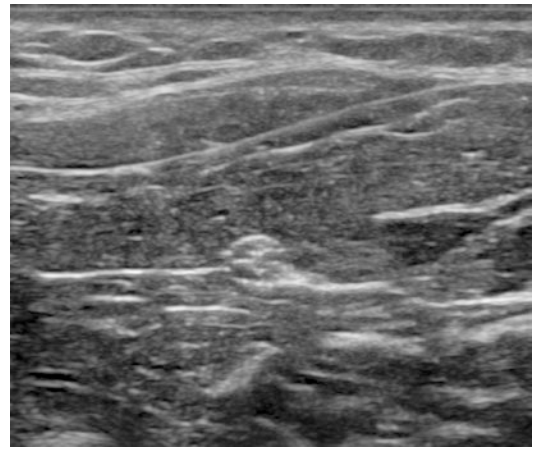

Fig. 1.17 Nerve echotexture. Transverse US image of the median nerve at the middle third of forearm. The nerve (white arrowhead) is characterized by a honeycombing appearance made of round hypoechoic areas in a homogeneous hyperechoic background

Anisotropy is a typical characteristic of tendon and refers to the sound reflection properties of tissue: tissues with low anisotropy tend to backscatter sound reflection, and those with high anisotropy

tend to reflect sound such that the angle of incidence equals the angle of reflection, making them much brighter with perpendicular insonation. So the angle of insonation does not alter the appearance of a nerve in the way that they alter the appearance of tendons. Tendons are more homogenously hyperechoic than nerves and have a distinctive fibrillary composition, and additionally, tendons, if followed proximally or distally, lead to muscle.

The shape of a nerve may also be different and vary between individuals, round, oval, triangular or irregularly shaped, which may change under compression by the probe or with the movement of a neighbouring muscle. Nerve may change its shape along its course, for example, from a triangular to a round cross section, or may present anatomic variants (e.g. bifid or trifid variants of the median nerve).

Knowledge of regional anatomy and topography is needed for the sonographic assessment of peripheral nerves and for localizing fine and deep nerves, and so characteristic anatomic reference points are used (often large vessels accompanying the nerves, which may be seen via Doppler imaging). The following anatomical landmarks are useful: the brachial artery for the median nerve in the upper arm, the superficial and deep flexor muscles for the median nerve in the forearm, the contents of the carpal tunnel for the median nerve at the wrist, the ulnar artery for the ulnar nerve at the wrist, the medial epicondyle for the ulnar nerve at the elbow, the radial groove for the radial nerve, the anterior and middle scalene muscles and the proximal subclavian artery for the brachial plexus, the popliteal artery for the distal sciatic nerve, the fibular head for the fibular nerve and the posterior tibial artery for the tibial nerve at the ankle.

Motor and motor-sensory nerves may be evaluated indirectly analysing the skeletal muscles which they innervate: we can evaluate muscular atrophy in case of chronic denervation as a decrease of the muscle's volume and fatty infiltration, which increases its echogenicity.

Ultrasound measurement of nerve size is very important because nerve enlargement is the most important diagnostic marker of an abnormal nerve: *cross-sectional area* and *swelling ratio* (the ratio between the cross-sectional area of the nerve at the site of maximal enlargement and that at an unaffected site) can be measured on transverse images, and diameter can be measured on longitudinal images.

For correct measurement, the transducer should be perpendicular to the nerve, with minimal pressure, and the site of maximal enlargement should be selected for the measurement of nerve size. Variability within a measurement can be reduced doing multiple measures. Measuring just inside the echogenic rim of the nerve is the preferred technique.

Placing the power Doppler box over the nerve and slowly increasing the gain can be useful to evaluate the vascularity of the peripheral nerves. No colour Doppler signal will be observed in the normal nerve.

Nerve mobility can be routinely assessed to exclude nerve entrapments.

1.3.2 Technique for Ultrasonographic Imaging of Peripheral Nerves

For the imaging of peripheral nerves, the patient lies with the 'region of interest' on the examination table.

Usually a linear probe with a frequency greater than 12–18 MHz is used. In the case of obese patients or the evaluation of deeply located nerves, a convex probe may be used, for a deeper penetration of the ultrasound waves.

For the evaluation of superficial nerves, using a thick layer of US gel or a stand-off pad can be helpful. In particular, such adjuncts are useful in the evaluation of fine nerves of the wrist.

Suggested Readings

Arda K, Ciledag N, Aktas E, Kadri Arıbas B, Köse K. Quantitative assessment of normal soft-tissue elasticity using shear-wave ultrasound elastography. AJR. 2011;197(3):532–6.

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