

Section I

Physics, safety, and patient handling

Chapter 1

Principles of ultrasound

JEREMY C. HEBDEN *and*
JANET M. RENNIE**Discovery of ultrasound**

The term ultrasound refers to sound with a frequency above that which can be detected by the human ear. The audible frequency range lies between 20 Hz and 20 kHz (one hertz equals one cycle per second, one kilohertz equals one thousand cycles per second), whereas the frequencies of sound waves used for diagnostic applications in medicine are of the order of one thousand times higher than this, with a range between 1 and 10 MHz (megahertz = one million hertz). Ultrasound imaging relies on the so-called pulse echo principle, which involves emitting a short burst of ultrasound and then listening for the returning “echo” after the sound has been reflected off appropriate surfaces. This is exactly the mechanism which has been employed by bats for millions of years to navigate their way around dark caves and to catch flying insects. Human interest in navigation using sound waves was significantly enhanced (if not initially inspired) by the sinking of the Titanic, which occurred when the ship collided with an iceberg in April 1912. Within a few years, ships were widely equipped with SONAR (Sound Navigation And Ranging) devices, which emit sound waves beneath the surface of the sea, and detect echoes from large objects within a radius of several miles. The technology advanced considerably during both world wars as it was utilized to detect submarines and mines. Ultrasound imaging of patients began to evolve in the late 1940s, and over the following decade simple (A-mode) systems were developed that could detect midline shift in head injury and the presence of foreign bodies in the orbit. These devices emitted ultrasound pulses and detected echoes along a single line through the tissue. To generate two-dimensional images, however, echoes must be acquired over multiple lines. Various methods of scanning the beam were explored, and Holmes [1] describes how some early imaging systems required subjects to be immersed in a water bath for the duration of the scan.

Reflection of sound from a moving object gives rise to a change in the observed frequency. This phenomenon, known as the Doppler effect, was first described by Austrian physicist

Christian Doppler in 1842. By measuring the change in frequency one can estimate the velocity of the moving object. That this is possible with some accuracy is well known to those who have been caught driving above the legal speed limit by police officers armed with a speed gun. Doppler measurement of blood flow has now become a standard feature of ultrasound systems, which commonly use color to display the velocity and direction of blood superimposed on conventional anatomical ultrasound images.

Ultrasound waves: basic principles

Diagnostic ultrasound uses a device known as a transducer, which can both emit and detect ultrasound waves. Transducers are made from a material known as a piezoelectric crystal, which vibrates millions of times per second when a short burst of electric current is applied to it. These crystals also have the property of detecting ultrasound by converting vibrations back into electrical energy. Piezoelectric properties occur in some naturally occurring crystals such as quartz, although most clinical transducers are based on a synthetic piezoelectric substance known as lead zirconate titanate (PZT). For imaging applications, the same crystal is used to transmit and receive, sending small bursts (pulses) of sound energy into the tissue and then “listening” for the returning echoes.

Ultrasound energy spreads by means of rapid alternate compression and expansion of the matter through which it travels, and therefore it cannot pass through a vacuum. Ultrasound passes through most biological tissues with a roughly constant speed (c) of 1540 metres per second. As shown in Fig. 1.1a, the distance between two consecutive peaks in an ultrasound wave is known as the wavelength (λ), and each pulse typically consists of a few wavelengths. The wavelength depends on the frequency (f) and speed of sound (c) according to the following equation:

$$c = \lambda f. \quad 1.1$$

The *intensity* of ultrasound is equal to the energy (in joules, J) transmitted each second per unit area, and is expressed in

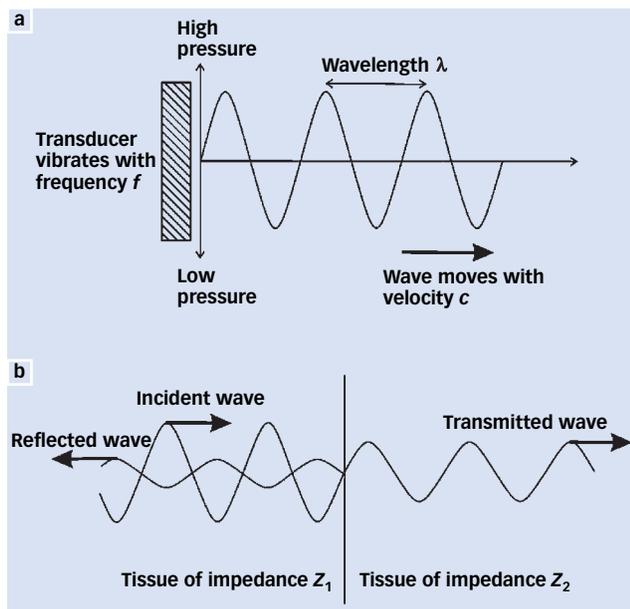


Fig. 1.1. (a) A pressure wave emitted by a continuously vibrating ultrasound transducer; (b) the partition of an incident wave into transmitted and reflected waves at a boundary between tissues of different acoustic impedance.

units of watts (W) per square metre ($1W = 1J$ per second). However, as ultrasound travels through tissue its intensity is steadily decreased as the sound energy is converted into heat. The attenuation causes the intensity I to decrease according to a so-called exponential function:

$$I = I_0 \exp(-\alpha x), \quad 1.2$$

where I_0 is the initial intensity, x is the distance traveled by the beam, and α is the *attenuation coefficient* of the tissue. The attenuation coefficient is strongly dependent on frequency, and rises roughly linearly between 1 MHz and 10 MHz.

When ultrasound passes through tissue, the most important interaction is reflection, which occurs when the beam encounters a boundary between two tissues having a different *acoustic impedance* (Z). This is illustrated in Fig. 1.1b. The greater the difference, the more the beam is reflected back towards the transducer. When a beam encounters bone, which has a much higher impedance than any soft tissue, almost all the ultrasound energy is reflected. Thus it is very difficult to view tissues directly beneath bones, and images display an artifact known as an acoustic shadow (see Fig. 4.23). Likewise, air has a much lower impedance than biological tissues, and virtually all ultrasound energy is reflected at an air/tissue boundary. Consequently it is necessary to use coupling gel to eliminate air between the transducer and the patient's skin during an ultrasound examination.

Other interactions between ultrasound and tissue include *refraction*, which occurs when the beam encounters a boundary between two tissues having a different speed of sound and *scatter*, which occurs when the beam encounters features with a size much smaller than the ultrasound wavelength.

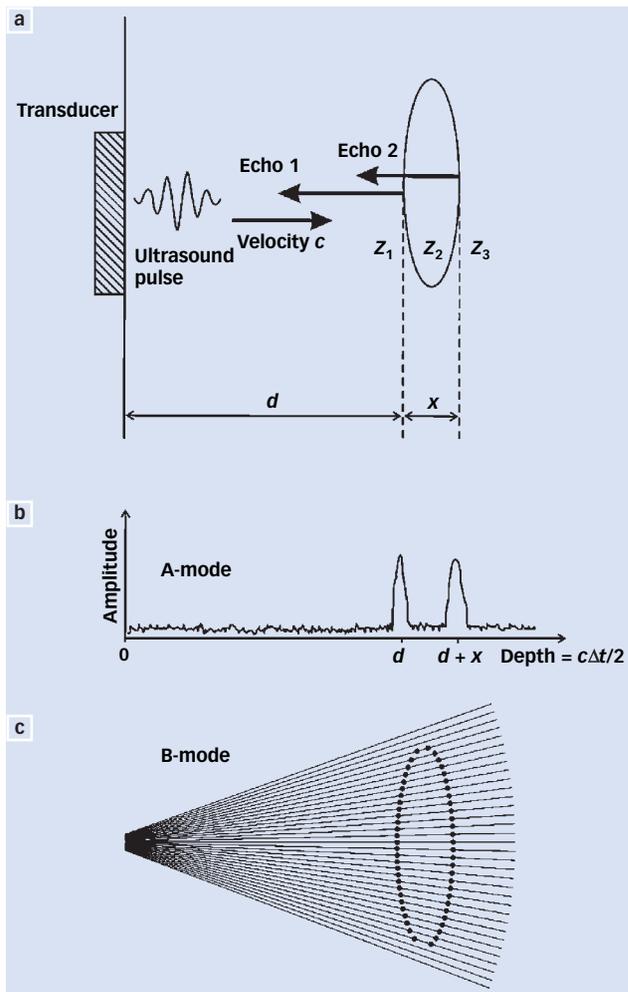


Fig. 1.2. (a) Reflected pulses of ultrasound (echoes) produced at boundaries between tissues of different acoustic impedance; (b) A-mode display of echo amplitudes as a function of depth below the surface; (c) B-mode display of echo amplitudes obtained from multiples lines through the tissue.

Further reading on the fundamental physics of ultrasound can be found in Hedrick, Hykes, and Starchman [2].

Echo location

The diagnostic use of ultrasound involves measuring and displaying the depths of boundaries between tissues of different acoustic impedance. This is achieved by determining the time taken (Δt) for the pulse to travel down to the boundary and for the echo to travel back up to the transducer (Fig. 1.2a). Since the velocity of sound (c) is fairly constant, the depth of the reflecting boundary (d) can be obtained from the following equation:

$$d = c \Delta t / 2. \quad 1.3$$

When a pulse of ultrasound energy is transmitted into tissue it is likely that several echoes will be detected at different

times, due to reflections occurring at different depths within the beam. The maximum depth that can be interrogated will depend on the time available before the next pulse is emitted.

Methods of displaying located echo information

A-mode

The earliest form of diagnostic ultrasound involved displaying echo strength as a function of depth along a single “line of sight” through the tissue. A single fixed transducer was employed, and the echo amplitudes were displayed on the screen of a cathode ray oscilloscope. This was known as “amplitude mode” or “A-mode,” and is illustrated in Fig 1.2b. Among its first uses was the detection of midline shift. Although A-mode is rarely used now, many modern commercial systems include an option to display the signal strength along a selected line of sight as a function of time (known as “M-mode”).

B-mode

To produce a two-dimensional (2D) ultrasound image it is necessary to acquire signals along multiple lines of sight through the tissue. For each line, the detected echoes are converted into bright spots on a screen, whose brightness depends on the echo amplitude and whose position depends on the echo arrival time and the direction of the beam (Fig 1.2c). This is consequently known as “brightness mode” or “B-mode.” A full 2D gray scale picture is formed by displaying the spots acquired from a series of successive lines of sight simultaneously. B-mode imaging therefore involves sweeping the beam across a plane through the tissue. The earliest imaging systems required the beam to be swept manually, with the transducer mounted on a device which continuously recorded its position and orientation as it was moved back and forth by the operator. Thus the picture was built up gradually. Thereafter, mechanical scanning systems were developed which used electric motors to translate or rotate one or more transducers across the tissue surface. By displaying many images (or frames) per second, diagnostic ultrasound became a *real-time* imaging technique. Modern ultrasound systems scan the beam without moving parts by employing an array of small transducers. *Linear arrays* use a row of up to 300 transducers, which are pulsed in small groups to emit a beam that travels perpendicular to the array. By selecting different groups, the beam is translated back and forth, sampling a rectangular slice across the tissue. *Phased arrays* consist of a much smaller row of transducers that are pulsed (almost) together. By introducing slight delays between the pulsing of neighboring transducers within the array, the beam can be steered, covering a sector of a circle. The angle of the sector depends on the scan depth and the number of distinct lines sampled by the scanner. The number of pulses generated per second (equal to the number of lines of sight sampled per second) is known as the pulse repetition frequency (PRF). Because of the finite speed of sound (c), the PRF limits the maximum depth from which echoes can be detected (d_{\max}):

$$\text{PRF} = c/2d_{\max}. \quad 1.4$$

The depth d_{\max} is therefore the depth of the displayed image. The maximum number of frames displayed per second (F) is given by the following equation:

$$F = \text{PRF}/N, \quad 1.5$$

where N is the number of sampled lines per frame. To avoid the perception of flicker, F is always more than about 20 frames per second. Higher values of F provide better images of rapidly moving objects such as heart valves. However, increasing F means increasing PRF or decreasing N , both of which mean reducing the size of the scanned area.

The small size (or “footprint”) of phased arrays makes them particularly suitable for neonatal examinations, where the curvature of the head makes it difficult to keep a larger array in contact.

3D imaging

Some commercial ultrasound imaging systems offer the facility to construct and display three-dimensional (3D) images. In principle, these enable the operator to determine spatial relationships in all three dimensions more accurately and more efficiently. There are two basic 3D scanning mechanisms. One involves mechanical scanning of a linear or phased array to acquire consecutive 2D slice images (which are then combined into a 3D image), while the other utilizes a static 2D transducer array that produces a beam which is steered electronically in three dimensions. Because of the significantly larger volume of tissue being scanned, 3D ultrasound imaging is inevitably slower than conventional 2D imaging. The principles and technology involved in 3D ultrasound imaging are described in detail by Fenster *et al.* [3].

Time-gain compensation

The attenuation of ultrasound as it passes through tissues means that echoes obtained from deep structures are much weaker than those obtained from more superficial tissues. Ultrasound intensity decreases exponentially with depth as described above. To compensate for this effect, the amplitudes of returning echoes are therefore multiplied by a number that *increases* exponentially with time. This has the effect of amplifying the echoes that originate at increasing depth. This is known as “time-gain compensation” (TGC), and allows similar features at different depths to give a similar appearance in the ultrasound image. The degree of compensation required (known as the “gain”) depends on the attenuation coefficient (α) of the tissue. Since α can vary depending on the tissue type, most ultrasound systems allow the operator to adjust the gain at different depths to give the best possible image. Some systems allow TGC to be estimated automatically according to the strength of the returning echoes.

Very weak echoes can be rejected entirely by setting an appropriate intensity threshold (before or after TGC is applied). This helps to suppress noise in the image. When setting up the sensitivity controls at the start of imaging, the

TGC and rejection threshold are best set to low levels. They can then be adjusted to give a suitable contrast for the main object of interest, and then refined to suppress unwanted small echoes. Many systems offer the ability to store the combinations of settings in the computer's memory which produce the best result for a given examination.

Resolution

The quality of an ultrasound image is determined by a broad variety of operational and systematic factors. The operator plays a vital role in choosing the best transducer for the job, in setting appropriate sensitivity and gain controls, and using coupling gel correctly. The image sector angle can be made narrower to increase the line density and/or the image depth. Overall performance can be checked with various commercially produced phantoms and test objects.

Spatial resolution is defined as the ability of an imaging system to distinguish between two closely spaced objects. For ultrasound imaging, this is characterized in terms of two independent measures: the axial resolution and the lateral resolution.

Axial resolution

Axial resolution refers to the ability of a system to distinguish between two objects that are separated by a small distance along the axis of the beam. In an ideal system the minimum resolving distance is equal to half the pulse length. Therefore, to achieve better axial resolution requires shorter pulses, which in turn requires smaller ultrasound wavelengths. The velocity of ultrasound in biological tissue is approximately constant, therefore to obtain a smaller wavelength requires higher frequency transducers. The wavelength of ultrasound transmitted through the body from a 1-MHz transducer is 1.5 mm, and from a 10-MHz transducer is 0.15 mm. However, in biological soft tissue the attenuation coefficient α is roughly proportional to the frequency, and thus higher frequency ultrasound is not able to penetrate as deeply as lower frequency ultrasound. In practice, therefore, the choice of probe is a compromise between the required depth of the image and the need for good spatial resolution. For imaging the neonatal brain a 7.5-MHz transducer is now usually chosen, and most of the pictures in this book were taken using a probe of this frequency. There are occasions in neonatal cranial ultrasound imaging when a 5- or 10-MHz transducer is required and most machines do offer a choice.

Lateral resolution

Lateral resolution is a measure of the ability to distinguish between two objects at the same depth. This is dependent on the width of the beam as it is scanned across the image plane. To distinguish between two objects they must not lie within the beam at the same time, and therefore their minimum separation is equal to the beam width. The narrowest region of the beam nearest the transducer where lateral resolution is good is termed the Fresnel zone, while the region further from the transducer where the beam broadens significantly is

known as the Fraunhofer zone. Lateral resolution within the Fresnel zone can be enhanced by focussing the beam, either by using an acoustic lens or by giving the transducer a concave surface. Lateral resolution is always worse than axial resolution. To achieve good lateral resolution it is essential to have a long Fresnel zone, which in turn needs higher frequencies. Again a compromise exists between requirements for high resolution and good penetration.

Contrast resolution

Contrast resolution is the ability of an ultrasound system to display different shades of gray corresponding to subtle changes in tissue reflectivity. This is largely determined by the energy of each pulse emitted into the tissue and by the amount of electronic noise contaminating the signals, although contrast resolution can be improved by averaging over signals from successive pulses emitted along the same line of sight. Contrast is also degraded by "speckle," which is a random pattern of bright and dark spots that occur when echoes from small scattering structures interfere with each other when they arrive at the transducer at the same time. Again, the effects of speckle can be significantly reduced by averaging successive images, although at a cost of reduced temporal resolution (see below).

The capability of the equipment as a whole must be considered; the measured signals undergo much processing before the image is presented on a video monitor. At this point the number of pixels, or dots, which make up the screen is also factor in the ability of a system to display small objects and distinguish them from each other. The quality of signal processing affects speckle reduction and interference from noise. In practice a good-quality 7.5-MHz imaging transducer operating on the infant head is capable of achieving a lateral resolution of about 1 or 2 mm, a performance a good deal worse than the limit imposed by the ultrasound wavelength alone. For comparison, magnetic resonance imaging or a good quality computerized tomography (CT) image with 512 pixels would achieve a resolution of about 0.5 mm.

Temporal resolution

Temporal resolution is defined as the ability of a system to distinguish between the times of occurrence of two separate events. Any tissue movement that occurs during the acquisition of a single ultrasound frame will appear blurred, and therefore to improve temporal resolution, the frame rate F must be increased. As described above, this requires the depth of the image and/or the number of lines in each frame to be reduced.

Doppler ultrasound

So far we have considered the behavior of ultrasound waves with regard to static boundaries within tissues. However, living tissues are not static, and some tissues, such as flowing blood, can move quite rapidly. In 1842 Christian Doppler first reported his observation that sound waves emitted by a source moving towards an observer have their wavelengths

compressed, which raises the pitch (frequency). Meanwhile, for a source moving away from the observer the wavelengths are stretched, which reduces the pitch. This “Doppler effect” is the cause of the familiar change in pitch of a train speeding through a station or of an ambulance siren as it passes by. The same phenomenon occurs with light waves, and is exploited by astronomers to determine the velocity of stars and galaxies relative to the Earth.

When ultrasound is reflected by moving tissues, a change occurs in the frequency of the reflected wave. A measurement of this change enables the velocity of moving tissues (and of flowing blood in particular) to be measured. In so-called color-flow Doppler imaging, blood flow information is superimposed in color onto the real-time gray-scale image.

The Doppler equation

When an ultrasound beam of transmitted frequency f encounters a target (such as a region of flowing blood) moving with a velocity v , the frequency of the reflected wave f_r is different by an amount f_d , given by the following equation:

$$f_d = (f - f_r) = f \frac{2v \cos \theta}{c}, \quad 1.6$$

where θ is the angle between the ultrasound beam and the direction of motion of the target and c is the velocity of sound in the tissue. Assuming both f and c are known and θ can be estimated, the velocity v can be obtained from a measurement of f_r . Estimating θ can be difficult, although when Doppler measurements of flow are combined with imaging, the orientation of the vessel of interest can be determined reasonably accurately, enabling good estimates of velocity to be obtained.

Continuous wave Doppler probes

A *continuous wave* (CW) Doppler probe consists of two transducer crystals: one to emit a continuous beam of ultrasound and one to detect the waves reflected back to the probe. The detected signals are amplified and filtered to reduce the effects of noise, and the shifts in the frequency are extracted by a process known as *demodulation*. For typical blood flow velocities and probe frequencies the Doppler shifts (f_d) happen to occur within the audible range of frequencies (i.e., a few kilohertz). Consequently they are commonly presented as an audio signal. The human ear is a sensitive instrument and the fact that Doppler signals can be heard in this way has undoubtedly helped the success of the method, as a trained operator can gain a great deal of information just by listening. Blood flowing in a vessel does not all flow at the same velocity and the changes induced by diseased or stenosed segments often produce a characteristic signal that can be detected by ear. However, a more objective analysis and the conversion of the information into a form that can be represented pictorially requires *spectral analysis*.

Pulsed Doppler instruments

A continuous beam of ultrasound does not allow the locations of reflecting structures to be determined, and therefore CW Doppler probes cannot identify the depths of blood vessels or

even establish if the acquired signal originates from more than one vessel within the beam. However, these limitations are overcome by pulsed Doppler instruments, which use a single transducer crystal to both transmit and detect. The Doppler shift frequency is determined by monitoring successive echoes obtained from reflecting structures at a given depth. If those structures move, those echoes will exhibit a small shift in their short waveform profile, from which the Doppler shift frequency can be derived. By selecting echoes arriving at a specific time following the emitted pulse, the instrument can extract the flow velocity information at a specific depth. Scanning the beam then allows a 2D map of flow information to be displayed. This Doppler map is normally combined with real-time B-mode imaging in a device sometimes known as a *duplex system*. This is a powerful investigative tool, as velocities can be sampled from known anatomical locations (see Fig. 1.3). The ability to view the orientation of blood vessels allows the velocities to be determined quite accurately.

Aliasing

Pulsed Doppler systems are only able to detect velocities unambiguously up to a finite maximum. The so-called sampling theorem [4] states that a continuous signal must be sampled at least twice at the highest frequency present in order to accurately recover the signal. This implies that to determine a Doppler shift frequency from a series of discrete measurements (i.e., from successive pulses) without ambiguity it is necessary that the PRF is at least twice that frequency. The maximum measurable frequency ($= \text{PRF}/2$) is known as the Nyquist frequency. If the Doppler shift frequency exceeds the Nyquist, then a phenomenon known as aliasing will occur, and the frequency is falsely recognized as being a lower frequency. The corresponding flow velocity, obtained by inserting the frequency into the Doppler equation, will be lower than its true value, and the flow direction can appear reversed (Fig. 1.4). Obviously if estimates of flow velocity are incorrect this can have negative clinical consequences.

A familiar example of aliasing is the curious apparent rotation of cartwheels recorded on film, which can often appear to be rotating much more slowly than they really are, or even in the reverse direction. This occurs when the film frame rate (the sampling rate) is insufficient to represent the true rotation rate of the wheel.

The example represented in Fig. 1.4 would be fairly easy to spot, but if the aliasing is very marked the Doppler signal can “wrap around” the forward and reverse channels making the resulting display appear like that of turbulent flow. If there is a possibility that aliasing is occurring, then this should be checked, either by using a CW Doppler probe (for which aliasing cannot occur), or by increasing the PRF. However, increasing the PRF decreases the maximum depth at which echoes can originate. As the maximum depth is reduced, it becomes increasingly feasible that late arriving echoes (from deeper structures) due to an emitted pulse can arrive during the measurement cycle for echoes produced by the following pulse (i.e., echoes are obtained simultaneously from two sampling volumes). The resulting ambiguity in sampling location can be

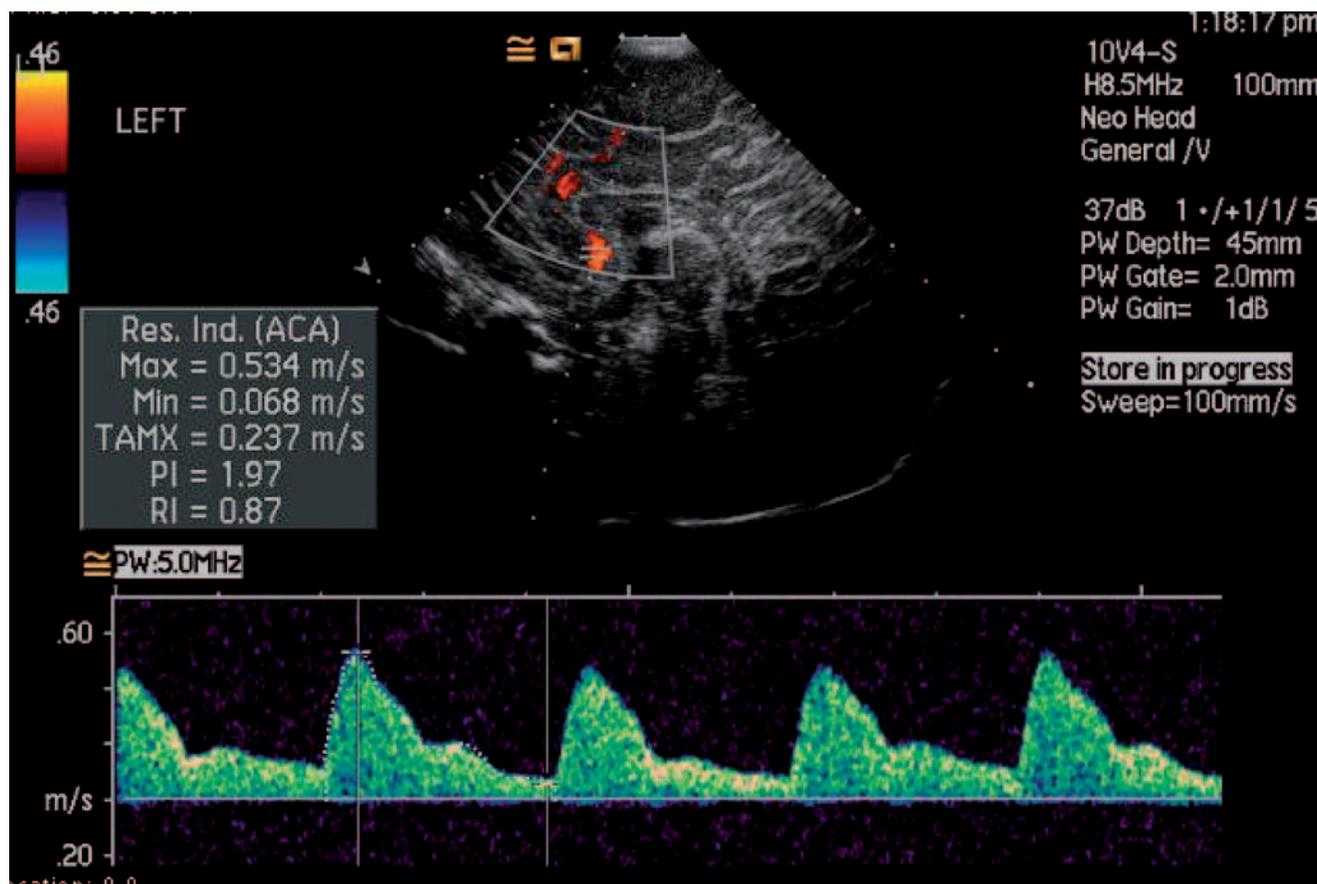


Fig. 1.3. Duplex Doppler output; system allows measurement of velocity information in a specified position within the image depth. Note display of the information as a sonogram after signal processing.

avoided if one of the potential sampling volumes is strategically placed within a region of no flow.

Doppler signal processing

The origin of the echoes used to determine blood flow is the back-scatter of the ultrasound by red blood cells. Since blood cells within a typical vessel are traveling with a broad range of velocities, echoes return with a range of Doppler-shifted frequencies. The complexity of the returned signal requires signal processing, both for audio interpretation and for visual presentation. Demodulation circuitry extracts the Doppler frequency signal, and amplifies it to an appropriate level. It is customary to subject the signal to a filter (known as a *wall-thump filter*) to remove a high-amplitude, low-frequency component due to reflection off the blood vessel walls. Wall thump filters are usually set for frequencies below 200 Hz, and therefore produce a gap in the velocity spectrum around zero. This can give rise to inaccuracies when distinguishing between minor degrees of reversed diastolic flow, and for neonatal cerebral work it is important to set the filter as low as possible. Further electronic circuitry is involved in distinguishing between forward and reverse flow, and in generating a sonogram.

A circuit known as a zero-crossing detector is sometimes used to quantify the principal Doppler frequency (and therefore the

dominant flow velocity), although its performance is unreliable when a large range of frequencies are involved or when the signal is particularly noisy. The best method of displaying and quantifying flow characteristics is spectral analysis. This involves performing a *Fourier transform* of the Doppler signals acquired over successive short windows of time (typically a few milliseconds). Although various display modes are used, the frequency content (i.e., the velocity content) is commonly exhibited as a function of time, with time updated continuously along the horizontal axis.

Color flow Doppler

The combination of pulsed Doppler and real-time imaging enables velocity information to be superimposed on the anatomical representation, and the concept of color coding the velocity information was first proposed in 1981 [5]. Real-time color flow mapping (CFM) systems are now commonplace. The usual pulse-echo B scan is used to provide a gray-scale image and multiple Doppler interrogations are performed in order to add information about the direction and velocity of blood flow within a selected region of the B scan.

Color flow mapping is perhaps of most use in cardiology, although the combination of Doppler and real-time imaging allows the location of vessels to be identified easily and significantly reduces the time required to examine flow in

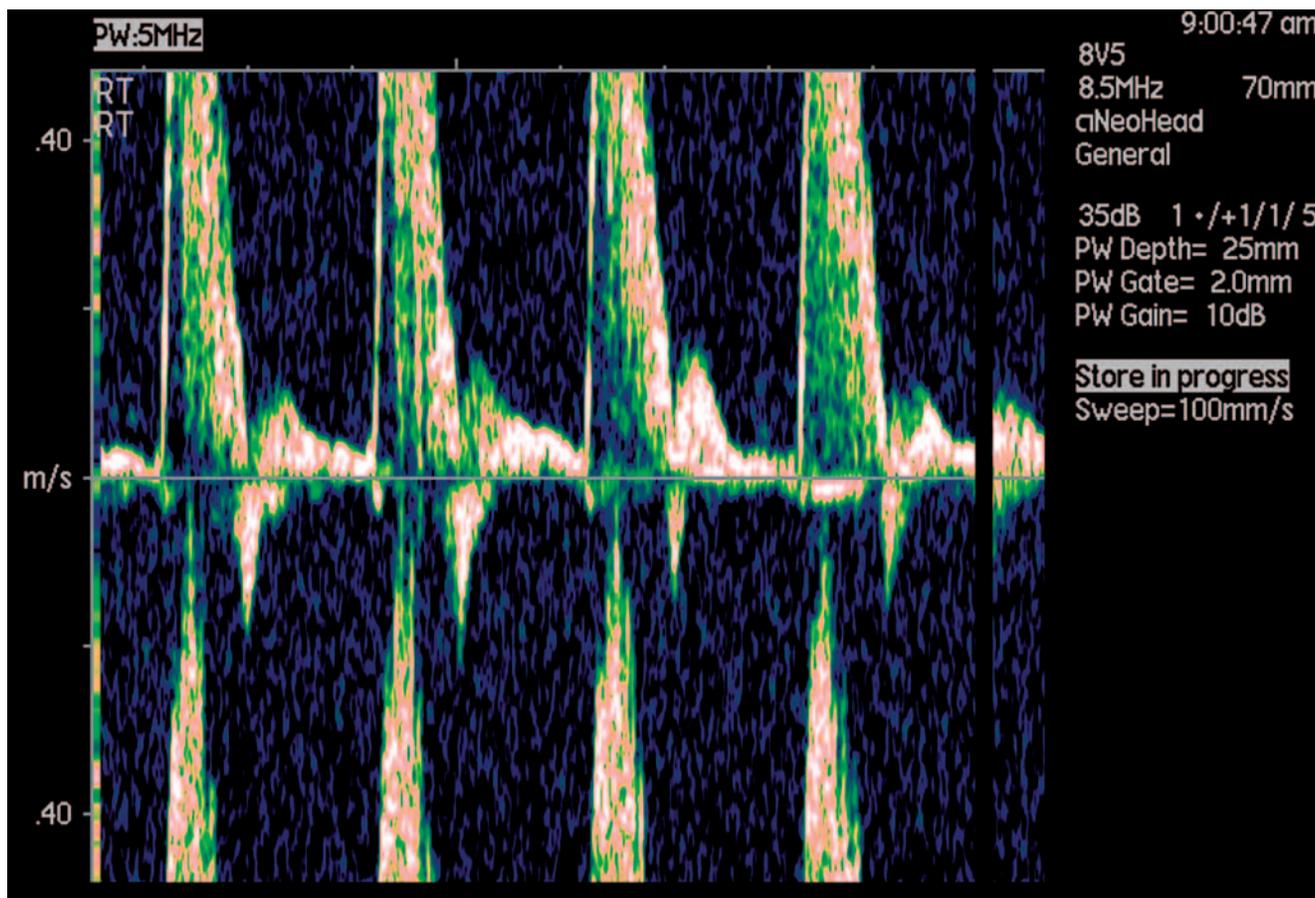


Fig. 1.4. Example of aliasing of range-gated Doppler velocity information. Tops of velocity profiles appear in the reverse channel.

neonatal cranial arteries (Fig. 1.5). Abnormal vascular leashes can be seen, allowing the diagnosis of arterio-venous malformations. Typically the velocities are displayed in shades of red through yellow for flow toward the transducer, and shades of blue through purple for flow away from the transducer. However, the scale is only approximately quantitative, mostly because of the averaging performed over a finite volume and period of time. Turbulent flow is typically represented as green, although color aliasing can occur with very high velocities in the same way as described for pulsed Doppler earlier. Aliasing can result in fast flow being coded as green (turbulent) when in fact the Doppler frequency has exceeded the Nyquist limit. Turbulent jets should therefore be checked, as advised for possible aliasing, with CW Doppler ultrasound.

Safety of ultrasound

Sound waves are a form of energy, and the amount of energy emitted by a typical ultrasound system is similar to that generated by the human voice during normal conversation. Nevertheless, exposure of tissue to very high levels of ultrasound is known to cause harmful biological effects. Given the possibility that diagnostic ultrasound may not be completely safe, regulatory authorities frequently examine the potential risks, and have occasionally issued guidelines for the prudent

use of ultrasound imaging. There is a general consensus that the potential for risk continues to grow as a result of the tendency for commercial scanners to provide an increasing level of exposure. This is a consequence of the manufacturers' quest to develop systems that deliver further improvements in diagnostic information and image quality. For example, achieving a higher spatial resolution requires higher frequencies, and therefore higher intensities to overcome the greater attenuation. Although a variety of potential harmful effects of diagnostic levels of ultrasound have been postulated, only two are generally considered to be of significant concern: thermal damage due to heating of tissue, and mechanical damage due to a phenomenon known as cavitation.

During a clinical ultrasound examination, almost all the energy in the ultrasound beam ends up being absorbed and therefore heats the tissue. The ability of the tissue to dissipate the heat depends on a variety of factors such as thermal conductivity and blood flow, but at typical diagnostic exposure levels the likely maximum rise in tissue temperature is generally considered too low to be a significant hazard.

Cavitation is the name given to the phenomenon that can cause sound waves to produce bubbles of gas in tissue. Tissues, like most liquids at normal pressures and temperatures, contain dissolved gas. During the negative pressure phase of an ultrasound wave, the dissolved gas can spontaneously come

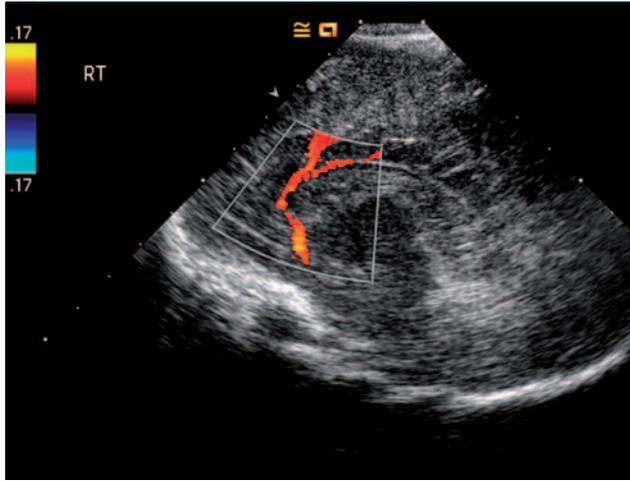


Fig. 1.5. Example of color flow mapping in the neonatal brain. The anterior cerebral artery appears red; flow towards the transducer.

out of solution producing microscopic bubbles. Once formed, these bubbles have a tendency to grow. If they reach a certain size they can vibrate in resonance with the ultrasound, producing violent forces within the immediate vicinity of the bubbles. This can lead to a phenomenon known as *microstreaming*, where high-velocity currents are established that are potentially damaging to biological molecules and structures. If the ultrasound intensity were to become sufficiently high (well above diagnostic levels) the bubbles can collapse destructively, producing exceedingly high local temperatures and pressures, leading to a variety of highly destructive effects.

As described by Wells [6], the authorities that regulate use of diagnostic ultrasound in some countries now require system manufacturers to provide an on-screen indication of the relative risk in terms of a “thermal index” (related to likelihood of thermal damage) and a “mechanical index” (related to risk of damage due to cavitation).

The overwhelming conclusion of the leading professional clinical and scientific organizations concerned with the use of the modality is that there are no confirmed harmful effects of diagnostic ultrasound. Nevertheless, it is recommended that the prudent operator will attempt to limit the exposure of his or her patient to a minimum, and a significant reduction can be achieved by following these simple suggestions (modified from Evans *et al.* [7], with permission):

1. Use the lowest transmitted power that will give a result.
2. Keep the duration of the examination to a minimum.
3. Use CW Doppler rather than pulsed wave if it will give a result.
4. Use the lowest PRF of pulsed Doppler that will allow the highest velocity to be measured.
5. Do not leave the Doppler beam irradiating a particular region for longer than is necessary. Switch back to imaging mode as soon as the Doppler examination is complete.
6. Use color Doppler only when necessary to make the diagnosis. Note that the Royal College of Radiologists [8] has produced specific guidelines on cranial ultrasound in infants that are available on their website: www.rcr.ac.uk/docs/radiology/pdf/ultrasound.pdf.

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Chapter 2 Principles of EEG

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Introduction

Electroencephalography (EEG) monitors the function of the neonatal brain and provides a sensitive real-time measure of cerebral activity. The neonatal brain is particularly vulnerable to changes in oxygenation and blood pressure and the effects of these physiological changes can often be detected by the EEG. In addition, it is now clear that clinical seizure expression in neonates is ambiguous and most seizures can only be detected by the EEG, which can also provide helpful information regarding the differential diagnosis (Chapter 7). Electroencephalography is also the most sensitive tool available for predicting neurodevelopmental outcome in neonates with early neonatal encephalopathy, particularly that due to hypoxic ischemia, and can provide this information much earlier than any other method. Many units use continuous amplitude integrated EEG (aEEG) to monitor cerebral activity. This provides a more limited measure of cerebral activity but nonetheless can be very useful, particularly if no other monitoring is available. Continuous monitoring with EEG is becoming a standard of care in many neonatal intensive care units (NICUs) around the world.

The aim of this chapter is to describe principles of EEG and aEEG recording. At the end of this chapter the reader should feel confident enough to record either the EEG or an aEEG from a newborn baby in the NICU. They will appreciate the characteristics of different recording devices, recognize the difficulties that are particular to the NICU environment which can impede recordings, recognize common sources of artifact that often mimic events such as seizures, and appreciate the difference between EEG and aEEG. Information on the appearances of the normal neonatal EEG and aEEG can be found in Chapter 6.

The electroencephalogram

The EEG is a measure of the electrical activity of the brain. Hans Berger first recorded this electrical activity from the scalp in humans in 1929. It is now known that EEG activity recorded from the scalp arises from ionic currents generated

by postsynaptic potentials caused by changes in the membrane permeability of dendrites and neuronal bodies. The electrical activity recorded from the scalp has a frequency range from 0.1 Hz to approximately 100 Hz and the amplitude is typically less than 200 μ V. Electrodes are used to make connections between the biological tissue (scalp generally in neonates) and a recording device (EEG machine). Unfortunately one electrode is not enough to measure the EEG; it has to be measured between two points on the scalp.

Technology of EEG recording

The EEG voltage recorded at the scalp is of low amplitude, being attenuated by the meninges, skull and scalp, and is of the order of microvolts (μ V; one millionth of a volt). The EEG activity is measured from metal electrodes that are fixed to the scalp using adhesive conductive paste and connected to a recording device that amplifies the tiny signals. Different regions of the brain generate different types of activity; therefore, electrodes have to be applied in a systematic fashion so that each area of the brain is studied.

The need for a general electrode placement format led the International Federation of Societies for Electroencephalography and Clinical Neurophysiology (IFSECN) to recommend a specific system of electrode placement for use in all laboratories under standard conditions. This is referred to as the 10–20 system of electrode placement. Specific measurements from bony landmarks (nasion, inion and left and right preauricular points) are used to determine the placement of electrodes. The term 10–20 is used because electrodes are placed at points 10% and 20% along lines between these bony landmarks (Fig. 2.1). The standard numbering system places even-numbered electrodes on the right side of the head and odd-numbered electrodes on the left with a letter designating the anatomical area. Midline electrodes are designated with the letter “z.” Therefore, an electrode on the left central region would be designated C₃; an electrode on the right parietal region, P₄; and an electrode on the mid frontal region, F_z (Fig. 2.1).

The way in which the EEG activity is displayed on the screen is dependent on a “montage.” A montage is a pattern of

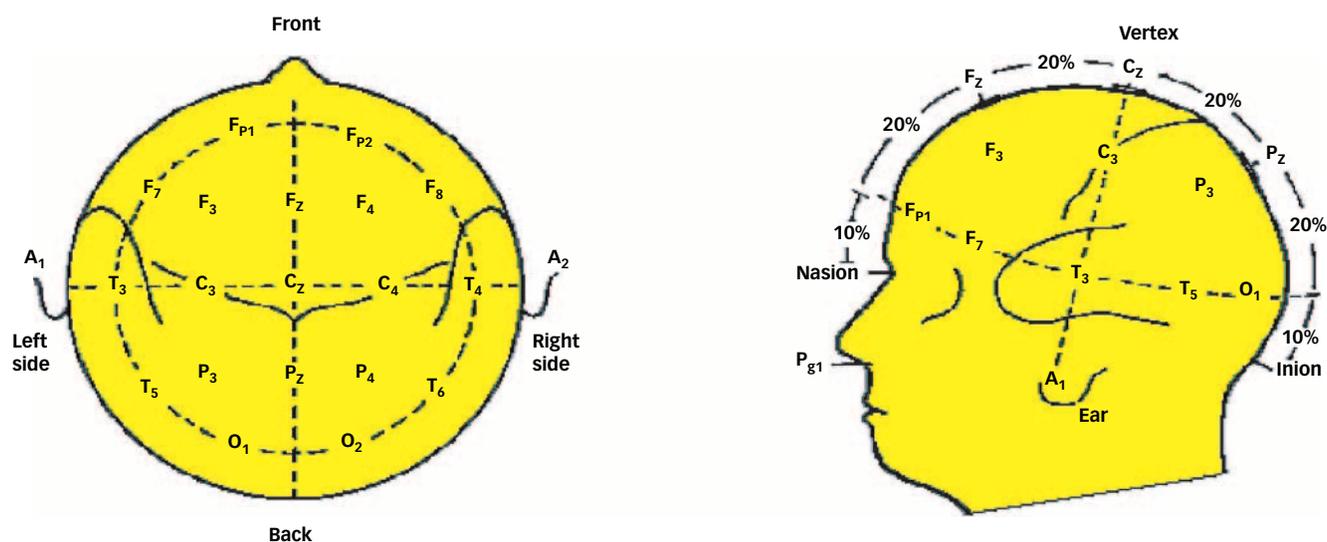


Fig. 2.1. The 10–20 system of electrode placement showing right and left designation and measurement landmarks. Fp, prefrontal; F, frontal; C, central; T, temporal; A, auricular; P, parietal; O, occipital.

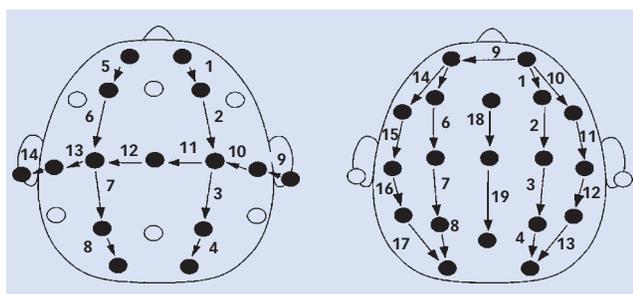


Fig. 2.2. Examples of EEG montages. This is the arrangement in which electrodes are linked for display. It makes EEG interpretation simpler.

electrode arrangement used for EEG display. Montages are generally selected so that recordings are made from rows of equidistant electrodes running from the front to the back of the head or transversely across it. Examples of montages are illustrated in Fig. 2.2.

Recording arrangements can be varied so that the potential difference is measured between pairs of scalp electrodes (bipolar) or between individual electrodes and a common reference point. In the latter arrangement the reference site can be either a relatively inactive site (such as linked ears) or a point connected to all the electrodes in use so that it reflects the average of the potentials at these electrodes. Most modern digital EEG machines use the common reference method for EEG signal acquisition, but have the ability to redisplay the EEG in any montage required termed “re-montaging.”

Recording the neonatal EEG

Recording the neonatal EEG requires certain modifications to allow for the fact that babies on the NICU are often very sick and incredibly small. The NICU environment is electrically very noisy and it can be difficult to extract small-amplitude EEG signals from a tiny baby who is often attached to lots of other monitoring equipment.

Full-term babies should ideally have a full set of electrodes applied to the scalp using the 10–20 system of electrode placement. However, it may not always be possible to apply a complete set of EEG electrodes to a premature or very sick unstable baby. The extra time involved in trying to achieve this is likely to worsen the baby’s condition. It is far more important to apply a fixed minimum number of electrodes as carefully and as efficiently as possible, ensuring that there is good contact and symmetry. Once a good recording has been obtained using these, it may be possible to apply more electrodes if required. Once electrode application has started there may be frequent interruptions for nursing/medical procedures such as suctioning and blood gas measurement. It can take over an hour to apply a complete set of electrodes carefully, particularly if the baby is ventilated. Quite often, electrodes become dislodged during the procedure and require re-application.

It is important to record from other non-cerebral channels during EEG monitoring. Electrocardiography or ECG is mandatory for neonatal EEG due to the frequent presence of ECG artifact, which may mimic stereotyped neonatal seizure patterns. There is also an increased risk of picking up pulse artifact through the fontanelles, which have not closed. Premature babies often have bradycardias and it is important to note this for optimum EEG interpretation. Any baby can have a bradycardia associated with a seizure. It is also important to record a respiration trace in the neonate because of the frequency of apneas during seizures, particularly in full-term babies. Rhythmic delta activity (slow wave activity) recorded in the neonatal EEG can often simply be respiration artifact since babies have high respiration rates of up to 100 breaths per min. Eye movement or electrooculogram (EOG) and muscle tone (EMG) channels are extremely useful to help determine sleep state in the baby.

Stockard-Pope *et al.* [1] in their excellent atlas of neonatal EEG recommend that at least eight scalp and two ear mastoid electrodes, covering all major areas of the brain, are used in