ULTRASOUND IMAGING

K2

K2.1 HISTORICAL REVIEW

Ultrasound refers to sound waves of frequencies higher than the hearing capacity of the human ear (between 20 Hz and 20 kHz). Frequencies used in medicine are usually greater than 2.5 MHz. Issuing from military and industrial applications for SONAR (sound navigation and ranging) and metal flaw detection, ultrasonic energy to "view" inside the human body was developed in parallel in hospitals of many countries around the world, including the **USA**, **Europe**, **Japan**, **China**, and **Australia.** The brief history below summarizes the main developments in the USA and Europe.

Ultrasound emerged in the **1940s** as a diagnostic, therapeutic, and then surgical tool using high intensities to break down tissue. The first diagnostic application is accepted to be due to **Karl Theodore Dussik** and collaborators in Austria. The work described in a paper published in **1942** showed images claimed to be of cerebral ventricles and included a discussion that low-intensity ultrasound could be used to locate brain tumors. Dussik and his collaborators had used a transmission method and the published images were later criticized as being artifacts due to reflections of sound waves within the skull. The transmission method of ultrasound imaging was subsequently dropped.

In the USA, **George Ludwig** applied ultrasound to systematic imaging of various tissues and organs from animals and measured the sound impedance of different types of gallstone, muscle, and fat, in the body. He was a naval officer and his work was classified and only released for publication in **1949**. At about the same time, **John Wild**, a surgeon, used the one-dimensional (A-mode) ultrasound investigation method to assess the thickness of bowel tissue under different surgical conditions. Then, with **John Reid**, a recent graduate in electrical engineering, they built a B-mode (two-dimensional) instrument able to visualize tumors by scanning breast lumps.

In **1953** in Sweden, **Carl Helmut Hertz**, a graduate student in physics (son of Gustav Ludwig Hertz in honor of whom the unit of frequency, in s^{-1}, was named) collaborated with cardiologist **Inge Edler** to make the first successful ultrasound measurement of heart activity, and generated the first echo-encephalogram (ultrasound brain scan). C. H. Hertz was already familiar with the application of ultrasound to non-destructive materials testing, and they made the measurements on an instrument borrowed from a ship construction company.

The work of obstetrician **Ian Donald** in Scotland led to the first diagnostic applications of ultrasonography. Again, by using borrowed industrial testing equipment, Donald and collaborators started by measuring the ultrasonic properties of various anatomical specimens. In **1958**, Donald, obstetrician **John MacVicar**, and physicist **Tom Brown**, after developing the equipment to diagnose pathological states in volunteer patients, published what many consider as the seminal paper on diagnostic medical ultrasonography (Fig. K2.1). Obstetric applications, including measurements to assess fetal size and growth, were developed by Donald and **James Willocks** and later refined by **Stuart Campbell**. And from the late **1960s**, ultrasonography became the method of choice to study pregnancy from start to birth and to diagnose its many complications.

In **1962**, **Joseph Holmes**, **William Wright**, and **Ralph Meyerdirk** developed the first compound contact B-mode scanner (which yields two-dimensional images). In the **late 1960s**, **Gene Strandness** and his team developed Doppler ultrasound as a diagnostic tool for vascular disease.

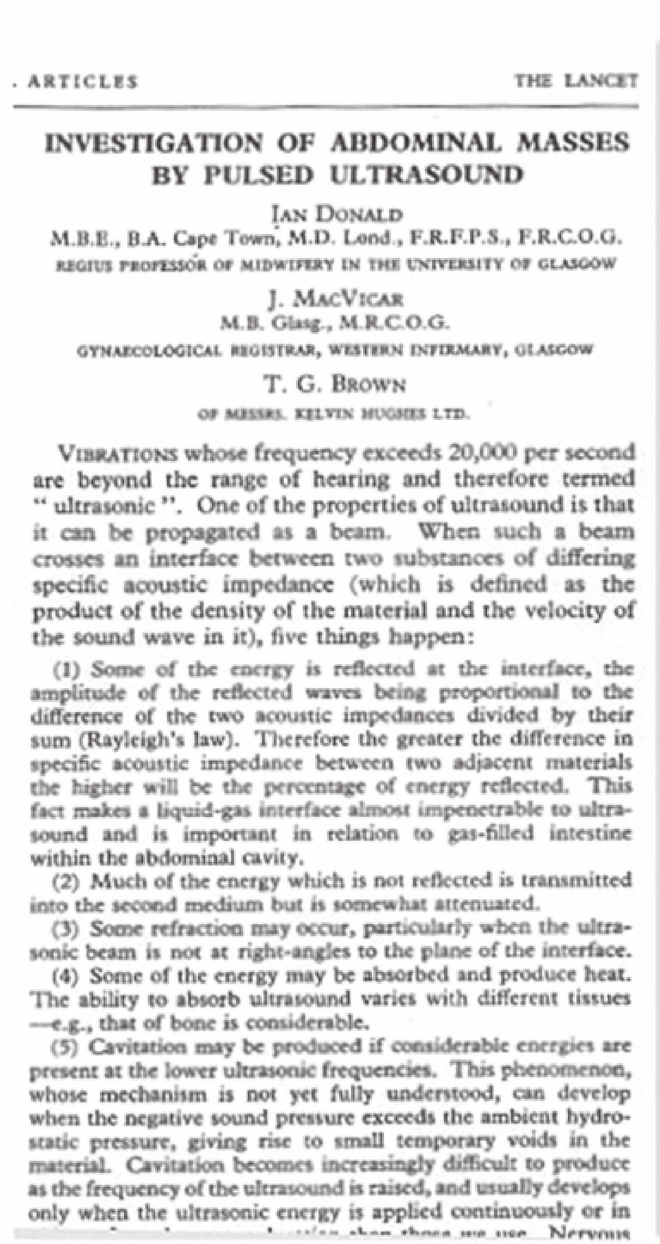
K2.2 ULTRASOUND WAVES

Sound waves of frequencies higher than 20 kHz, the upper limit that can be heard by the human ear, are called ultrasound. In contrast to electromagnetic radiation that can propagate in a vacuum, sound waves propagate in a material medium, inducing cyclical local modifications in density, pressure, and temperature. They are **longitudinal waves**, i.e., the modifications are along the direction of wave propagation – not perpendicular to it as would be the case for transverse waves (Fig. K2.2a). Ultrasound frequencies used in medical imaging are in the range ~1.5–7.5 MHz, with higher frequency images more suitable for more superficial tissues.

The speed of propagation of a wave, c (ms^{-1}), its frequency, f (s^{-1}), and wavelength, λ (m) are logically related by:

$$c = f\lambda \qquad \text{(K2.1)}$$

A typical value of c in tissue is ~1500 m s^{-1}; for a frequency of 1.5 MHz the wavelength is 1.0 mm.



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INVESTIGATION OF ABDOMINAL MASSES BY PULSED ULTRASOUND

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VIBRATIONS whose frequency exceeds 20,000 per second are beyond the range of hearing and therefore termed "ultrasonic". One of the properties of ultrasound is that it can be propagated as a beam. When such a beam crosses an interface between two substances of differing specific acoustic impedance (which is defined as the product of the density of the material and the velocity of the sound wave in it), five things happen:

(1) Some of the energy is reflected at the interface, the amplitude of the reflected waves being proportional to the difference of the two acoustic impedances divided by their sum (Rayleigh's law). Therefore the greater the difference in specific acoustic impedance between two adjacent materials the higher will be the percentage of energy reflected. This fact makes a liquid-gas interface almost impenetrable to ultrasound and is important in relation to gas-filled intestine within the abdominal cavity.

(2) Much of the energy which is not reflected is transmitted into the second medium but is somewhat attenuated.

(3) Some refraction may occur, particularly when the ultrasonic beam is not at right-angles to the plane of the interface.

(4) Some of the energy may be absorbed and produce heat. The ability to absorb ultrasound varies with different tissues —e.g., that of bone is considerable.

(5) Cavitation may be produced if considerable energies are present at the lower ultrasonic frequencies. This phenomenon, whose mechanism is not yet fully understood, can develop when the negative sound pressure exceeds the ambient hydrostatic pressure, giving rise to small temporary voids in the material. Cavitation becomes increasingly difficult to produce as the frequency of the ultrasound is raised, and usually develops only when the ultrasonic energy is applied continuously or in

Fig. K2.1 First page of the paper considered to be one of the most important in medical imaging: **Donald**, I., **MacVicar**, J., and **Brown**, T. G. (**1958**) Investigation of abdominal masses by pulsed ultrasound. *Lancet*, **1,** 1188–1195.

The amplitude A of a wave corresponds to the maximum displacement from equilibrium of the particles composing the medium induced during the wave cycle (Fig. K2.2). The energy of the wave is proportional to A^2 (see Chapter A3 for a more detailed treatment of waves).

The speed of sound c, in a medium, depends on the average local density ρ (in units of $\mathrm{g\,m^{-3}}$ in Table K2.1), and compressibility κ ($\mathrm{m\,g^{-1}\,s^{2}}$) (Comment K2.1) according to:

$$c = \sqrt{\frac{1}{\kappa\rho}} \tag{K2.2}$$

The **acoustic impedance** Z, the square root of the density/compressibility ratio, is a physical property of the medium.

$$Z = \sqrt{\frac{\rho}{\kappa}} \tag{K2.3}$$

Combining Eqs (K2.2) and (K2.3):

$$Z = \rho c \tag{K2.4}$$

where c is the speed of sound in the medium (Comment K2.2).

Formally, Z is defined as the ratio of the pressure in the sound wave at a given point to the particle velocity at that point. Its units are ($\mathrm{g\,cm^{-2}\,s^{-1}}$). As is seen in Comment K2.2 and below, it is a useful parameter to describe the behavior of an acoustic wave as it traverses different media.

Acoustic impedance values are also useful for describing the pressure amplitude and intensity ratios between the reflected and transmitted signals and the incident intensity (p_r/p_i, p_t/p_i, I_r/I_i, I_t/I_i, respectively).

The strongest reflected signal is obtained when the incident beam is at right angles to the interface (incident angle equal to zero in Fig. K2.3). The reflected beam is **back-scattered**. In this case the ratios are given by:

$$R_P = \frac{p_r}{p_i} = \frac{(Z_2 - Z_1)}{(Z_1 + Z_2)} \qquad T_P = \frac{p_t}{p_i} = \frac{2Z_2}{(Z_1 + Z_2)} \tag{K2.5}$$

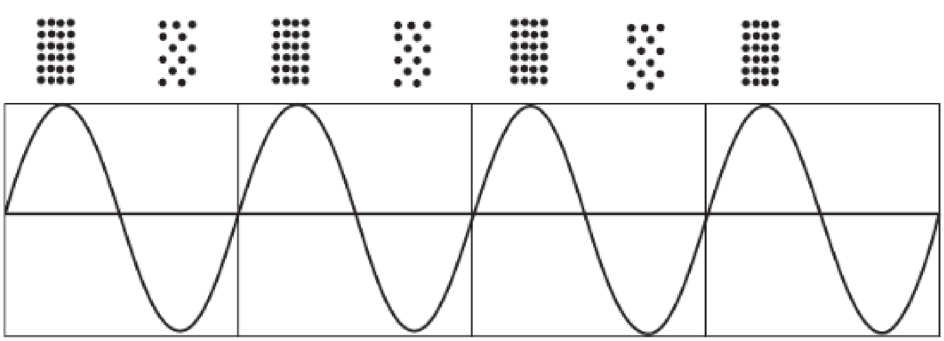

Fig. K2.2 Longitudinal sound waves are pressure waves in air.

TABLE K2.1 ACOUSTIC PARAMETERS AND THEIR UNITS FOR DIFFERENT TISSUE TYPES (ADAPTED FROM SMITH AND WEBB, 2010)

Medium	$Z\,(10^{-5}\,\mathrm{g\,cm^{-2}\,s^{-1}})$	$c\,(\mathrm{m\,s^{-1}})$	$\rho\,(\mathrm{g\,m^{-3}})$	$\kappa\,(10^{-11}\,\mathrm{cm\,g^{-1}\,s^{2}})$
Air	0.00043	330	1.3	70 000
Fat	1.38	1450	925	5.0
Brain	1.58	1540	1025	4.2
Blood	1.59	1570	1060	4.0
Kidney	1.62	1560	1040	4.0
Liver	1.65	1570	1050	3.9
Muscle	1.7	1590	1075	3.7
Bone	7.8	4000	1908	0.3

COMMENT K2.1 BIOLOGIST'S BOX: COMPRESSIBILITY

The compressibility of a medium is defined as the relative volume change when pressure is applied.

$$\kappa = \frac{1}{V}\left(\frac{\partial V}{\partial p}\right)$$

In thermodynamics, the value of κ depends on whether the change is observed at constant temperature or constant entropy. In the case of a solid medium, these two values are very close to each other. The units of compressibility are inverse pressure. Qualitatively, the compressibility of a medium is inversely proportional to the forces that maintain its average structure; the softer the medium, the higher its compressibility.

$$R_I = \frac{I_r}{I_i} = \frac{(Z_2 - Z_1)^2}{(Z_1 + Z_2)^2}$$
$$T_I = \frac{I_t}{I_i} = \frac{4Z_1Z_2}{(Z_1 + Z_2)^2} \qquad \text{(K2.6)}$$

where the reflection and transmission pressure coefficients (R_p, T_p) and reflection and transmission intensity coefficients (R_I, T_I) are related by:

$$T_p - R_p = 1$$
$$T_I + R_I = 1 \qquad \text{(K2.7)}$$

The intensity of a wave is the square of its amplitude. The amplitude of a sound wave corresponds to the maximum pressure in the cycle, so that

$$R_I = |R_p|^2$$

The reflected signal intensity is proportionally stronger with the square of the difference between the impedance values (Eq. (K2.6)). The extreme case is when one of the Z values is zero. But then the T ratio will be zero; the wave will not penetrate to deeper structures! The opposite extreme is when two tissue types have equal impedances, $Z_1 = Z_2$. But then there will be no acoustic interface between them and no reflected beam ($R_I = 0$).

The image is formed by the reflected waves from acoustic boundaries between different tissue types. It is, nevertheless, important to have good transmitted intensity in

COMMENT K2.2 REFRACTIVE INDEX AND ACOUSTIC IMPEDANCE

The refractive index n of a medium is a measure of how much the speed of light in a vacuum c_{LV} is decreased when it propagates in the medium:

$$v = c_{LV}/n$$

The change of direction (refraction) of a light wave as it crosses between media of different refractive index (n_1, n_2) is governed by Snell's Law:

$$n_1 \sin\theta_i = n_2 \sin\theta_t$$

which can be written

$$\frac{1}{v_1}\sin\theta_i = \frac{1}{v_2}\sin\theta_t$$

where θ_i and θ_t are the angles of incidence and transmission (refraction) of the wave, respectively (Fig. K2.3).

A similar equation holds for an acoustic wave.

$$\frac{1}{c_1}\sin\theta_i = \frac{1}{c_2}\sin\theta_t$$

where c_1 and c_2 are the speed of sound, respectively, in medium 1 and medium 2.

Applying Eq. (K2.4)

$$\frac{\rho_1}{Z_1}\sin\theta_i = \frac{\rho_2}{Z_2}\sin\theta_t$$

The refractive index analogy for sound is the ratio of density to acoustic impedance.

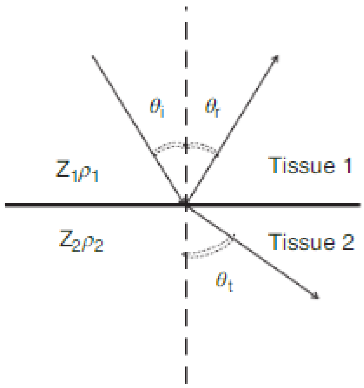

Fig. K2.3 Snell's law diagram

> **COMMENT K2.3 WATTS, DECIBELS AND BEAM ATTENUATION**
>
> The intensity of the ultrasound beam is given in Watts cm^{-2}, units of energy per area.
>
> The ratio between the attenuated intensity and the incident intensity can be expressed in decibels; the decibel is a logarithmic expression of an intensity ratio.
>
> The attenuation coefficient in units of length^{-1} is defined by Eq. (K2.8). The conversion to decibel units is given by:
>
> $$\mu(\text{dB cm}^{-1}) = 10 \times \log_{10}(e)\mu(\text{cm}^{-1})$$
>
> $$\log_{10}(e) = 0.4343 \text{ so that } \mu(\text{dB cm}^{-1}) = 4.343\,\mu(\text{cm}^{-1}).$$
>
> A useful rule-of-thumb for the interpretation of Eq. (K2.8) is that each 3 dB reduction corresponds to a reduction of a factor of 2 in intensity.

order for the sound waves to penetrate to the deeper tissue layers (Section K2.4).

Intensity and intensity ratio are measured in units of Watts cm^{-2} and decibels, respectively (Comment K2.3). Values of acoustic impedance and speed of sound in different tissue types are shown in Table K2.1, together with corresponding density and compressibility.

According to Eq. (K2.2), sound waves travel faster in more rigid (less compressible) and/or the less dense media of propagation. In bone, even though it is about twice as dense, sound travels almost three times faster than in other tissues (Table K2.1) because of its very low compressibility (a factor of 13 lower than for blood).

The values for soft tissue in Table K2.1 are very similar. In practice, this indicates that for ultrasound waves only a small fraction of the wave intensity will be reflected at an acoustic interface between two tissue types. At a fat–muscle interface, for example, $Z_1 \sim 1.38$ and $Z_2 \sim 1.7$ in the units of Table K2.1. Substituting in Eq. (K2.5), $R_I = 1\%$; $T_I = 99.9\%$; most of the beam is transmitted to deeper tissue. The imaging information is contained in the 1% of the incident intensity that is reflected (see Section K2.4).

For a wave that crosses a boundary between soft tissue ($Z_1 \sim 1.6$) and bone ($Z_2 \sim 7.8$), $R_I = 43.5\%$; $T_I = 56.5\%$; the reflected signal is strong but less than 60% of the beam is transmitted to deeper tissue.

K2.3 HEALTH PHYSICS, ABSORPTION, AND ATTENUATION OF ULTRASOUND WAVES IN BIOLOGICAL TISSUE

K2.3.1 Effects of Ultrasound

Biological effects of sound waves have only been observed in animals for extremely high incident intensities. During medical examinations, the absorbed energy is small and no ill effects on human health have been observed. It is nevertheless recommended not to use high intensities in prenatal examinations.

Thermal effects. Negligible heat is produced in tissue by ultrasound waves during usual imaging procedures. More heat is produced in Doppler mode (Section K2.4.2), in which it is recommended to use the minimum power for a useful signal and to limit exposure time. Local heating effects at higher intensities are used in ultrasound heat therapy to relieve pain.

Mechanical effects of ultrasound include **cavitation** (the production of microscopic bubbles in the tissue), which has been applied in the development of contrast agents for ultrasound imaging. High-intensity ultrasound has therapeutic applications; for example, to break up deposits such as gall or kidney stones or to ablate tumors or specific tissue in **focused ultrasound surgery (FUS)**.

K2.3.2 Attenuation

Ultrasound beam intensity is reduced as it penetrates into tissue through successive reflections at acoustic boundaries, and attenuation by **scattering** and **absorption** or energy loss through production of heat. Note that a relatively new technique, HIFU (high-intensity focused ultrasound), makes use of this heat to thermally ablate tumors.

As in the case of radiation (Eq. (K1.3)), the attenuation of a sound wave follows an exponential law. At a depth z (cm), the incident intensity I_0 is reduced to I_z

$$I_z = I_0 \exp(-\mu z) \tag{K2.8}$$

where μ (cm^{-1}) is the linear attenuation coefficient.

It turns out that the attenuation coefficient in tissue is approximately linear with frequency with a value of **~1 dB MHz^{-1} cm^{-1}** for soft tissue. The decibel unit (dB) and its relation to μ are defined in Comment K2.3. In-depth exploration becomes more difficult the higher the

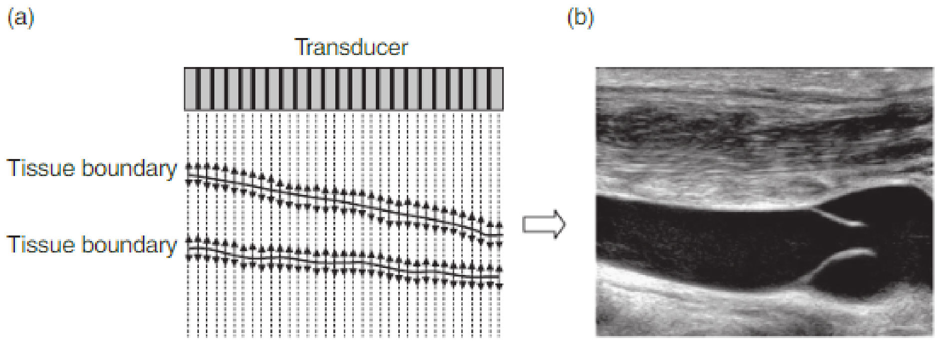

Fig. K2.4 Principles of ultrasound imaging. **(a)** At boundaries between tissues, a small fraction of the energy of pressure waves from a transducer is scattered back to the transducer, where it is detected. The depth of the boundary is detected from the speed of sound in the tissue. **(b)** The image formed by electronic steering of the beam. The intensity in each pixel is proportional to the back-scattered signal. (From Smith and Webb, 2010, with permission.)

frequency, and ultrasound imaging instruments apply corrections to compensate for the attenuated signal as a function of depth.

As in the general case for the diffraction of waves by different-sized objects (Chapter G1), sound waves are scattered by structures of smaller size than the wavelength. In soft tissue ($c \sim 1500\,\mathrm{ms}^{-1}$) the wavelength is 1 mm for a frequency of 1.5 MHz (Eq. (K2.1)). Much smaller particles, such as red blood cells, will act as point objects and scatter isotropically. As for the general radiation case, there will also be interference between waves scattered by neighboring particles. Back-scattering will be constructive for particles that are close together. Analogously to the formation of diffraction fringes, interference will be constructive or destructive as a function of angle for particles that are further apart, resulting in a complicated intensity pattern called **speckle**, which contributes to image noise.

K2.4 PRINCIPLES OF ULTRASOUND IMAGING

The basic principle of ultrasound imaging is illustrated in Fig. K2.4. Ultrasound waves are sent through the tissue under examination. The back-scattered reflected waves (the "echoes") at acoustic boundaries are detected as a function of time, so that the depth of the boundary can be calculated from the knowledge of the speed of sound through the tissue. The image is formed by the successive lines of reflected intensity as the beam sweeps the sample. Image intensity is proportional to the intensity of the echo.

K2.4.1 Imaging Mode

The instrumental setup for ultrasound imaging is illustrated in Fig. K2.5. A frequency generator produces short periodic voltage pulses that are amplified and transmitted to a transducer, which converts them into a sound wave that is transmitted into the body. The transducer both transmits the high-power pressure wave pulses and detects the low-intensity echo signals. It is, therefore, important to keep them apart. This is done by a transmit/receive switch, which opens sequentially to allow passage either from the frequency generator to the transducer or back from the transducer to the imaging line. The signals from back-scattered pressure waves are analyzed according to the time they reached the transducer and processed electronically to yield a real-time image display on a monitor.

The **transducer** contains a piezoelectric element that produces the incident pressure waves when it receives an electric signal, and in the inverse process transforms

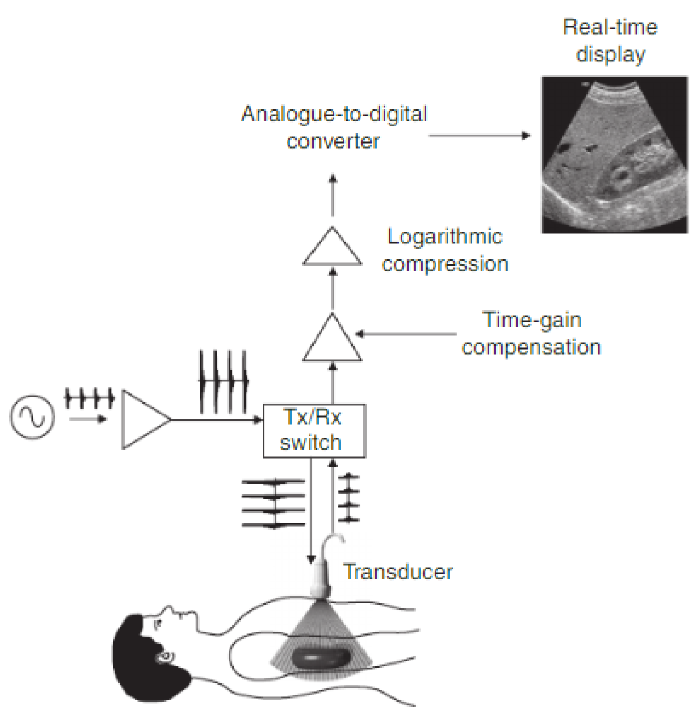

Fig. K2.5 Ultrasound imaging setup. (From Smith and Webb, 2010, with permission.).

the echo pressure waves into electric signals. A **damping layer** absorbs the waves scattered backward from the piezoelectric element. A **matching layer** at the external face increases the efficiency of the setup significantly by providing acoustic coupling between the element and the patient, through a zone of intermediate acoustic impedance.

$$Z_{matchinglayer} = \sqrt{Z_{piezoelectric} Z_{skin}} \quad \text{(K2.9)}$$

The matching layer is also a **quarter-wavelength plate** – its thickness equal to one-quarter the ultrasound wavelength to maximize energy transmission in both directions.

Transducers have **large bandwidths** around a specified central frequency – e.g., ±2 MHz for a central frequency of 3 MHz. This means that a single transducer can be used for many applications.

In single-element instruments, the transducer is mounted on a rotating or oscillating system to allow sectorial scanning (Fig. K2.6a).

Transducers in recent instruments, however, contain **large arrays** of piezoelectric elements that enable the acquisition of two-dimensional images while the transducer is held in a fixed position (Fig. K2.6b). The array can be linear (for a rectangular image) or convex (for a sectorial image), with each element generating an image line.

Phased arrays are rectangular probes with which a sectorial scan is obtained by applying voltage pulses to each element at slightly different times (out of phase). The sum of waves produces an effective wave front that can be focused and steered within the body by appropriate phasing. Phased array probes have the advantage of being able to scan a large area with a relatively small probe – e.g., for exploration via an intercostal pathway.

Lateral resolution, focusing, and axial resolution (Fig. K2.7). **Lateral resolution** describes the minimum

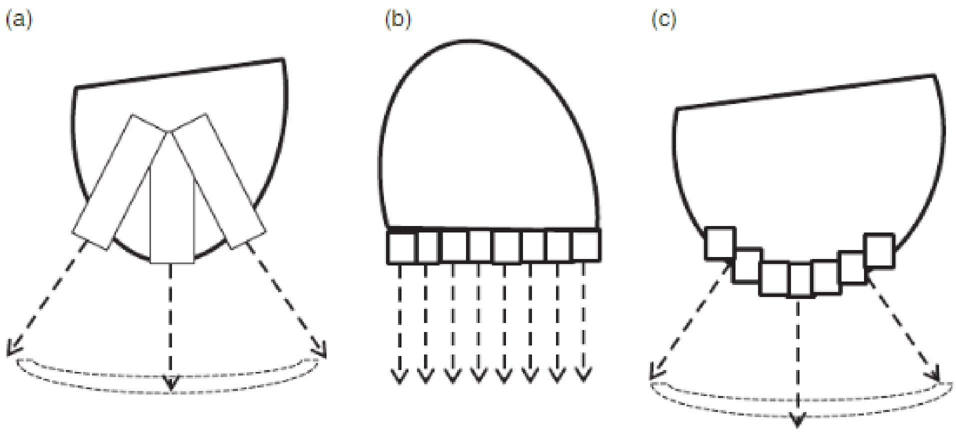

Fig. K2.6 Transducers (schematic). (a) Single-element rotating or oscillating system to allow sectorial scanning. (b) Fixed large array of piezoelectric elements (planar). (c) Fixed large array (curved).

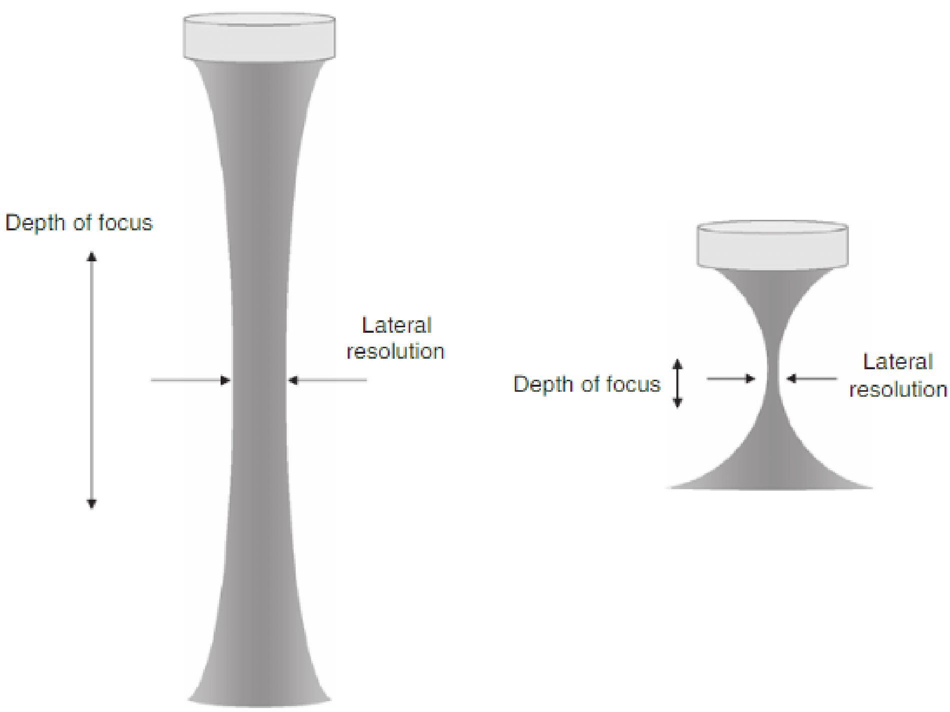

Fig. K2.7 Lateral resolution, focusing and axial resolution. (From Smith and Webb, 2010, with permission.)

distance for objects at the same depth to be distinguished as separate. Lateral resolution is determined by the width of the ultrasound beam as it diverges from the transducer. **Beam focusing** reduces beam width and concentrates ultrasound energy in a certain zone, the focal point; the finer the focused beam, however, the shorter the depth of focus. In transducer array instruments, lateral resolution will also depend on the number of beams and their spacing. Focusing is obtained by using a curved plastic lens (similarly to the optical case) or by electronic means in phased arrays (see above).

Axial resolution describes the minimum distance for two acoustic boundaries along the beam propagation line to be distinguished as separate. Recall that depth is determined from the time it takes the echo wave to reach back to the transducer. Axial resolution will, therefore, depend on the vibration frequency of the probe and quality of the damping layer, since the probe should have stopped vibrating after detecting an echo before being able to detect a subsequent one. In principle, axial resolution cannot be better than one-half the wavelength of the sound wave.

Contrast resolution describes the capability to distinguish between structures whose acoustic impedance is close. Current transducers are sensitive to 150 dB and can detect echoes of weak intensity. Microbubbles of gas in a lipid or protein shell of diameter 2–10 μm are used as **contrast agents**. The highly compressible gas has an enhanced acoustic response compared to surrounding tissue or blood that reflects back not only the main frequency component (f_o) of the incident ultrasound beam, but also higher harmonics ($2f_o$, $3f_o$, etc.). The presence of strong harmonics makes it possible to distinguish the echoes from microbubbles from those from other tissue boundaries. The transducer bandwidth should be sufficiently broad to emit the (f_o) incident beam and to detect the higher harmonic echoes.

A-mode, M-Mode, and B-Mode Scanning

B-mode or **brightness scanning** is the most common ultrasound procedure. It produces a two-dimensional sectorial or rectangular (depending on transducer geometry) image of the tissue cross-section examined. The image is built up of echo lines from the transducer array or by mechanically scanning a single transducer plotted as a function of depth (obtained from the time between incident pulse emission and echo reception). Brightness is proportional to the intensity of the echo.

A one-dimensional image is obtained from an **amplitude (A-) mode scan**. The line-image is a plot of the back-scattered amplitude as a function of time, which provides the depth from which the echo originated. A major application of A-mode scanning is to measure corneal thickness by applying a small probe directly on the center of the previously anesthetized eyeball.

In an **M-mode** or **motion scan**, a series of A-mode lines, representing different depths, is displayed as a function of time, with the brightness of the lines proportional to the amplitude of the echo. A straight line represents a motionless boundary. Pulses will appear on the line when an acoustic boundary moves perpendicular to the ultrasound beam direction (the surface of the heart, for example). M-mode scans are used in echography of the heart and the fetal heart.

Correcting for Beam Geometry and Other Artifacts

Lobes develop on either side of the divergent main ultrasound beam from interference effects between wavelets as they leave the transducer. The side beams will reflect back off acoustic barriers, outside the region studied, to create artifacts in the image called **clutter**. Clutter will also arise from other undesired interference effects, e.g., from multiple reflections. Electronic processing of the signals can reduce this. Recall from Section A3.3.2 that signals in the time and frequency domains, respectively, are related by Fourier transformation. Multiplication in one domain corresponds to convolution in the other. A time window introduced in the time domain to eliminate echoes from side lobes will correspond to a convolution of all echo frequencies by the Fourier transform of the time window and will act as an appropriate filter. Clutter, mainly due to the main frequency reflections, will also be reduced by **harmonic imaging** (see above), in which the second harmonic of the echo is analyzed. In the absence of contrast agents, the second harmonic is quite weak but the analysis profits from much better lateral resolution and reduced clutter.

The depth dependence of attenuation is discussed in Section K2.3.2. It leads to a wide dynamic range in the received signal, with, for example, strong echoes from boundaries with large differences in acoustic impedance close to the surface and much weaker ones from boundaries between two types of soft tissue deeper down. This effect is corrected by instrument electronics in a process called **time-gain compensation**. Echo signals are not amplified by the same factor but according to the time received. Early echoes from shallow structures are amplified by a smaller factor than those received later from deeper structures.

Speckle (see Section K2.3.2), which produces a grainy appearance, is the main artifact in ultrasound imaging. It is reduced by **compound scanning**, an approach in which the ultrasound image is acquired from multiple angles. Speckle is due to interference effects, which are angle dependent. When the "views" from various angles are combined to form the final image, "true" features will appear in most of them, while the speckle pattern will depend on angle and be reduced by averaging out. Compound scanning is also useful in image boundaries of irregular curvature that are parallel to the beam and in reducing image artifacts due to **acoustic enhancement** when the beam crosses a volume of exceptionally low attenuation (such as water-containing cysts), or the opposite effect, **shadowing**, in which structures are "hidden" behind a strongly attenuating medium (such as bone or a solid tumor).

K2.4.2 Doppler Mode

A classic illustration of the Doppler effect is the rising frequency of sound from a fast-approaching train that goes through a station without stopping and the falling frequency after the train goes past the platform and recedes into the distance. It is as if the wave is pushed "compressed" by the approaching speed and expanded as the source moves away. The same principle applies to sound reflected off a moving object. Blood flow in Doppler mode ultrasound is measured by reflecting ultrasound waves off red blood cells. The effective velocity v of the red blood cells is obtained from the frequency shift Δf of the echo with respect to the incident frequency. The Doppler effect is highest when the incident beam is parallel to the direction of motion and null when it is perpendicular to the direction of motion. The shift is given by Eq. (K2.10), in which the cosine term determines the component of the incident beam that is parallel to the motion.

$$\Delta f = f_R - f_0 = \frac{2f_0 v \cos\theta}{c} \tag{K2.10}$$

In the case of blood flowing in a direction roughly parallel to the transducer, the signal will be very low, taking up small positive and negative values around $\Delta f = 0$ ($\cos 90^\circ = 0$). In **power Doppler** mode, the intensity of the signal is integrated, with both positive and negative amplitudes contributing as the squares. The advantages of the method are its higher sensitivity and that artifacts due to high flow rates are eliminated. A disadvantage is that information on the directional of flow is lost.

A Doppler mode measurement is illustrated schematically in Fig. K2.8.

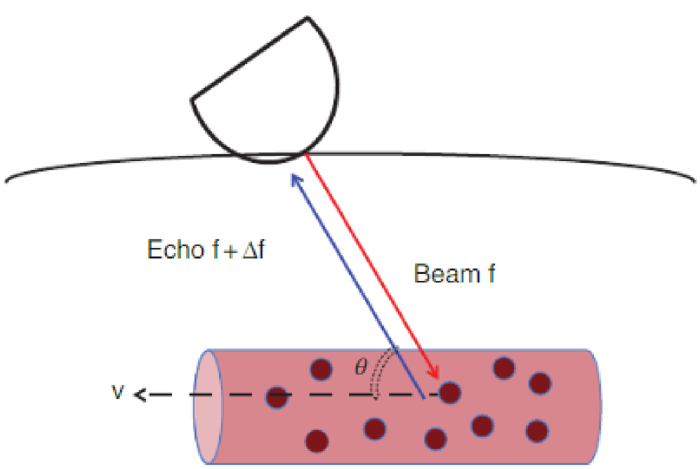

Fig.K2.8 Ultrasound Doppler measurement. The beam of frequency *f* is in red, the echo of frequency *f*+Δ*f*, from the particle moving with velocity *v*, is in blue.

In **continuous mode**, the transducer array is divided into two sections, one that emits a continuous incident beam and one that receives the Doppler shifted echo. The measurement area is the overlap of the areas delimited by the two beams, with no depth resolution. The measured speed is the average speed of blood flow in the measurement area.

In **pulsed mode**, the transducer alternates between emitting the ultrasound beam and receiving the echo. An important advantage of pulsed wave Doppler is good spatial resolution that enables focusing on a specific blood vessel, by combining with a B-mode image (duplex image). Results are plotted with frequency on the *y*-axis, time on the *x*-axis, and brightness proportional to the density of red blood cells contributing to the particular frequency. There is a range of velocities in a blood vessel that changes significantly over the cardiac cycle. Fourier transform analysis splits the signal into its component frequencies to inform on the velocities present. By using fast Fourier transform methods, the analysis is performed and plotted on the screen in real time in addition to the duplex image (triplex image). In **color flow** imaging, the Doppler information is overlaid on the B-mode image. Blue and red represent, respectively, flow toward or away from the transducer, with color intensity proportional to flow velocity. Frequency dispersion can also be coded in color, usually in green or yellow.

Similarly to B-mode imaging, the incident frequency must be adapted to the exam. Low frequencies (2–5 MHz) are used for deep vascular exploration (e.g., in the abdomen), while higher frequencies (7.5–10 MHz) are required to examine superficial blood vessels. Bandwidth settings should also be adapted to the velocity range. Pulses beyond the high end of the range will create artifacts due to *aliasing* (an *alias* pulse that appears at a "wrong" frequency within the range). Pulses corresponding to slow velocities will be missed if outside the low end of the range. Finally, the θ angle between the incident beam and the vessel (Eq. (K2.10)) should be kept below about 40° for good signal sensitivity ($\cos 40^\circ = 0.766$).

K2.5 CHECKLIST OF KEY IDEAS

- Ultrasound refers to sound waves of frequencies higher than the hearing capacity of the human ear (between 20 Hz and 20 kHz). Frequencies used in medicine are usually greater than 2.5 MHz.
- Sound waves are longitudinal pressure waves.
- A typical value for the speed of sound in tissue is ~1500 m s^{-1}; for a frequency of 1.5 MHz, the wavelength is 1.0 mm.
- The compressibility of a medium is the relative volume change when pressure is applied.

- The acoustic impedance is the square root of the density/compressibility ratio. It is a physical property of the medium.
- Sound propagation across boundaries follows Snell's law. The refractive index analogy for sound is the ratio of density to acoustic impedance.
- The strongest reflected signal is obtained when the incident beam is at right angles to the interface and the reflected beam is back-scattered. It is proportional to the square of the difference between the acoustic impedance values.
- The image is formed by the reflected waves from acoustic boundaries between different tissue types.
- Sound waves travel faster in more rigid (less compressible) and/or less dense media of propagation.
- The absorbed energy is small during medical examinations and no ill effects on human health have been observed. It is nevertheless recommended not to use high intensities in prenatal examinations.
- Thermal effects are negligible, except in the case of high-intensity focused ultrasound, a technique that is used for the thermal ablation of tumors.
- Mechanical effects include cavitation, which has been applied in the development of contrast agents for ultrasound imaging. High-intensity ultrasound has therapeutic applications, for example to break up deposits such as gall or kidney stones or to ablate tumors or specific tissue in focused ultrasound surgery (FUS).
- The attenuation coefficient in tissue is approximately linear with frequency, with a value of ~1dB MHz^{-1} cm^{-1} for soft tissue. In-depth exploration becomes more difficult at higher frequency and ultrasound imaging instruments apply corrections to compensate for the attenuated signal as a function of depth.
- Sound waves are diffracted by structures of smaller size than the wavelength (1 mm in soft tissue for ~1500 ms^{-1} speed of sound and a frequency of 1.5 MHz). Much smaller particles such as red blood cells will act as point objects and scatter isotropically.
- In imaging mode, echoes at acoustic boundaries are detected as a function of time. The depth of the boundary is calculated from the knowledge of the speed of sound through the tissue. The image is formed by the successive lines of reflected intensity, as the beam sweeps the sample. Image intensity is proportional to the intensity of the echo.
- The transducer contains a piezoelectric element that produces the incident pressure waves when it receives an electric signal, and in the inverse process transforms the echo pressure waves into electric signals.
- Lateral resolution describes the minimum distance for objects at the same depth to be distinguished as separate. It is determined by the width of the ultrasound beam as it diverges out from the transducer.
- Beam focusing reduces beam width and concentrates ultrasound energy at a focal point; the finer the focused beam, however, the shorter the depth of focus.
- Axial resolution describes the minimum distance for two acoustic boundaries along the beam propagation line to be distinguished as separate.
- Contrast resolution describes the capability to distinguish between structures whose acoustic impedance is close.
- B-mode, or brightness scanning, is the most common ultrasound procedure. It produces a two-dimensional sectorial or rectangular (depending on transducer geometry) image of the tissue cross-section examined.
- An amplitude or A-mode scan produces a one-dimensional line-plot of the back-scattered amplitude as a function of time, which provides the depth from which the echo originated.
- In an M-mode or motion scan, a series of A-mode lines representing different depths is displayed as a function of time, with the brightness of the lines proportional to the amplitude of the echo.
- Instruments include electronic signal processing to correct for beam geometry and other image artifacts.
- Compound scanning, an approach in which the ultrasound image is acquired from multiple angles, is used to correct for speckle.
- In Doppler mode, blood flow is measured by reflecting ultrasound waves off red blood cells. The effective velocity of the red blood cells is obtained from the frequency shift of the echo with respect to the incident frequency.
- In continuous Doppler mode, the transducer array is divided into two sections, one that emits a continuous incident beam and one that receives the Doppler shifted echo. There is no depth resolution. The measured speed is the average speed of blood flow in the measurement area.
- In pulsed Doppler mode, the transducer alternates between emitting the ultrasound beam and receiving the echo. An important advantage of pulsed wave Doppler is good spatial resolution that enables focusing on a specific blood vessel, by combining with a B-mode image.

Suggestion for Further Reading

Barrie Smith, N., and Webb, A. (2010). *Introduction to Medical Imaging: Physics, Engineering and Clinical Applications*. Cambridge: Cambridge University Press.