Physical principles of medical ultrasound

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Sound and ultrasound

Acoustics is the scientific field that studies sound. Sound is a form of mechanical periodic molecular displacement (vibration) of matter. The time it takes for a vibration cycle to complete is called a period. The number of vibration cycles that occur during a set time is referred to as the frequency. The frequency *f* of a vibration is the inverse of its period *T*:

$$f = \frac{1}{T}. \qquad \textbf{Eq. 1}$$

Sound with frequencies below 20 cycles per second, *i.e.*, below 20 Hz, is called infrasound. Although infrasound is too low to be heard by human beings, it can be perceived (felt).

The audible range is defined by frequencies between 20 Hz and 20,000 Hz (20 kHz). This range has been defined by the average hearing of healthy 18-years-old men. Frequencies higher than 20 kHz are referred to as ultrasound.

Figure 1 shows some clinical application of ultrasonics and their respective frequency bands.

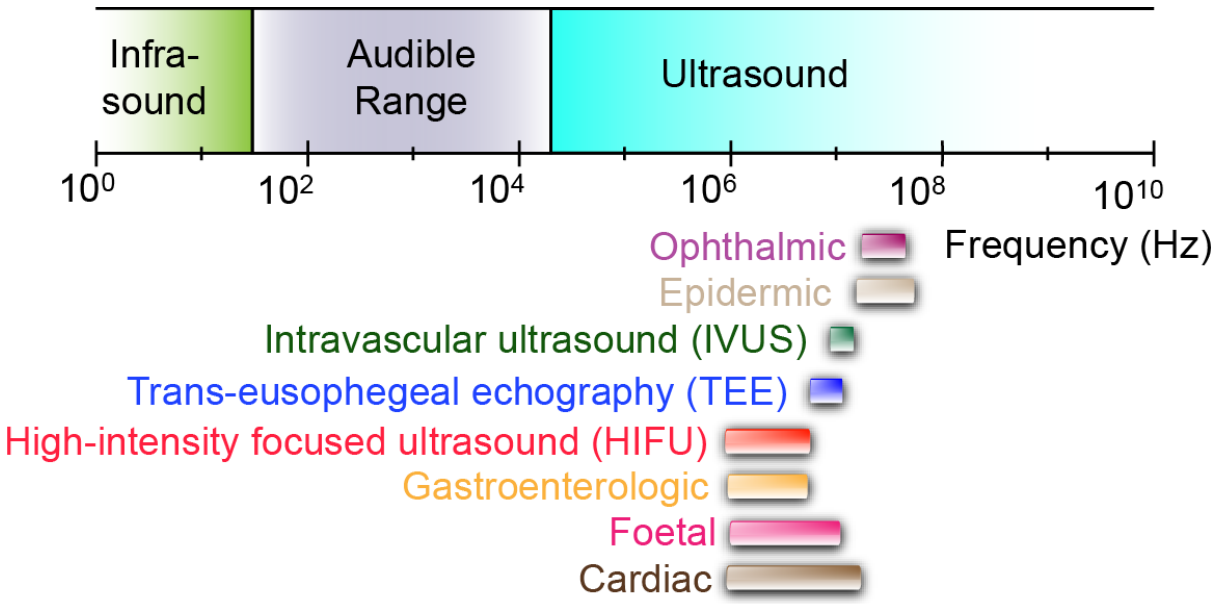

Figure 1 – Clinical applications of ultrasound and their corresponding frequency bands.

Sound propagates from a source through matter. Although many different elastic wave types exist in solid materials, fluids only support longitudinal waves. Longitudinal waves, also known as acoustic waves, displace matter only in the direction on propagation (*cf*. Figure 2). As human tissue consists mostly of fluid materials, primarily longitudinal waves are generated and observed in the field of medical ultrasonics.

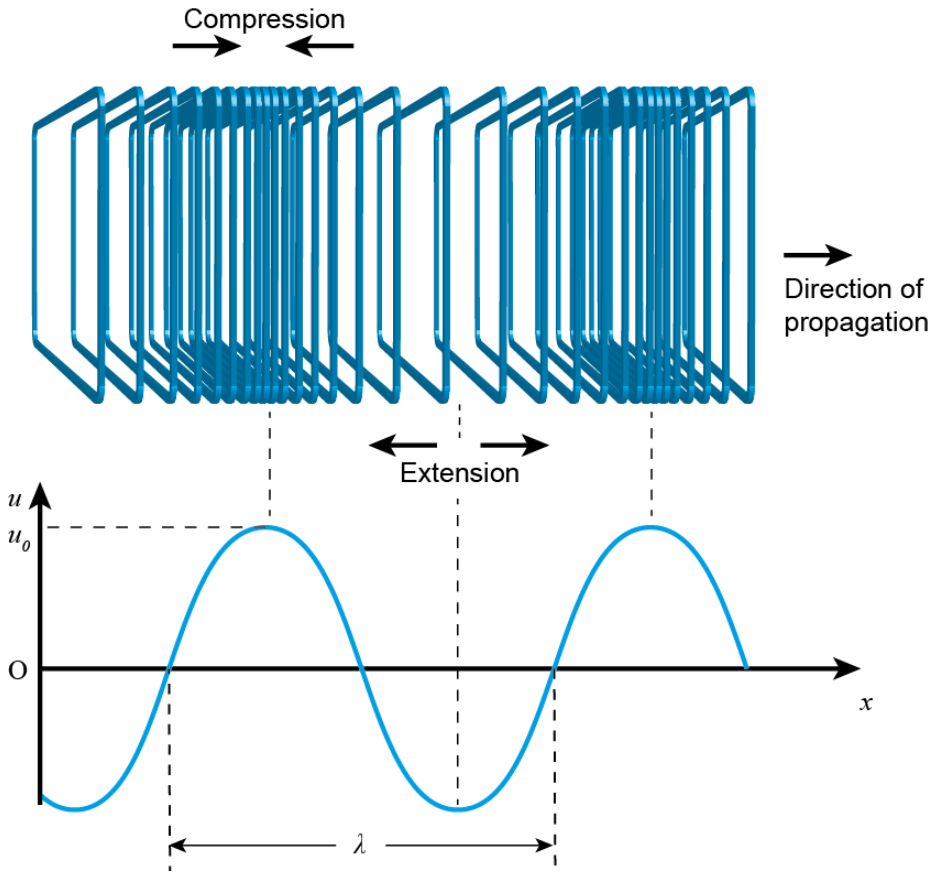

Figure 2 - Schematic representation of the axial displacement of matter by a longitudinal sound wave.

The highest displacement in a sound wave is called the displacement amplitude. Generally, the matter displacement in space and time by a low-amplitude sound wave has the form

$$u(x,t) = u_0 \sin 2\pi \left(\frac{t}{T} - \frac{x}{\lambda} \right), \qquad \textbf{Eq. 2}$$

where u_0 is the displacement amplitude and λ is the wavelength of the sound (*cf.* Figure 2).

Notice the minus between $\frac{t}{T}$ and $\frac{x}{\lambda}$ in (2): obviously, the wave at given time farther from the source is equal to the wave at earlier time closer to the source. Taking only one dimension into account, the compressive and extensive displacements are related to local pressure changes by the equation of motion from which the wave equation is derived (*cf.* Appendix, Eq. A 1 – A 3).

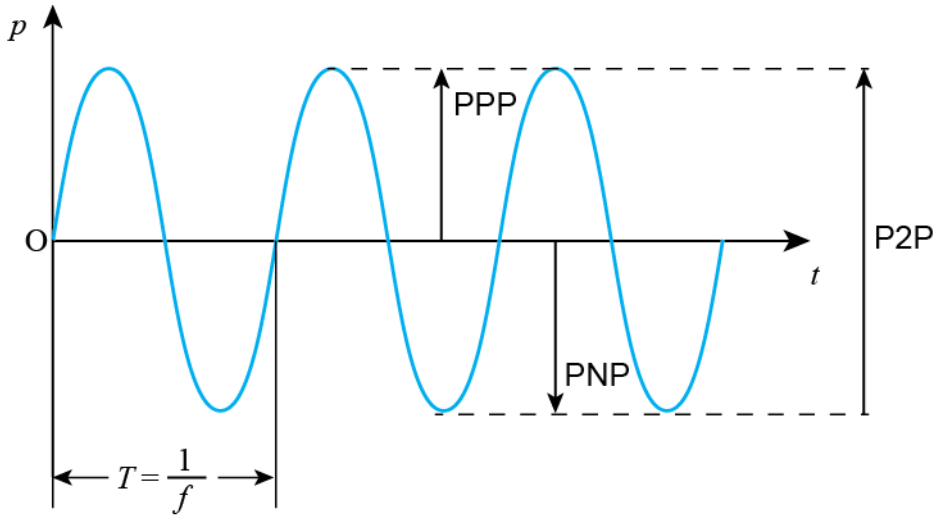

Figure 3 – Common parameters used to express pressure amplitudes: peak-positive or peak-compression pressure (PPP), peak-negative or peak-rarefaction pressure (PNP), peak-to-peak pressure (P2P).

Figure 3 shows some often-used parameters to express the pressure amplitudes of medical ultrasound, which are handy especially if the sound waves are asymmetric.

Let us define an imaginary sound source with power W, *i.e.*, every second, a certain amount of energy is radiated from the source. The power is an intrinsic property. At equal distances from the source, we can define a surface S, through which this energy must pass. The power per unit surface area is called the instantaneous intensity:

$$I = \frac{W}{S}. \qquad \textbf{Eq. 3}$$

The averaged derived intensity of a harmonic sound wave at a point in a sound field is:

$$< I >= \frac{{p_{\mathrm{A}}}^2}{2\rho c}, \qquad \textbf{Eq. 4}$$

where p_{A} is the pressure amplitude, c is the speed of sound in the medium, and ρ is the density of the medium. Thus, for a point source, the surface through which the energy must pass is a sphere of radius r (*cf.* Figure 4) and a surface area $S = 4\pi r^2$. Consequently, for a point source, the intensity is inversely proportional to the

distance to the source squared, and the acoustic pressure is inversely proportional to the distance itself. This acoustic pressure decay with distance is called geometric damping.

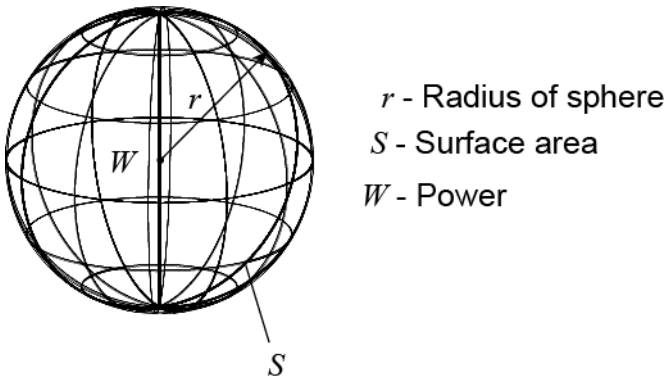

Figure 4 – Radiated field through a spherical surface *S* at a distance *r* from a point source with power *W.*

Thermal and viscous material properties are other causes of damping of the acoustic wave (*cf*. Figure 5). Damping coefficients are frequency-dependent. In human tissue, the damping coefficient is proportional to the frequency to a power between 1.0 and 1.4. Thus, the higher the frequency, the lower the penetration depth of the sound.

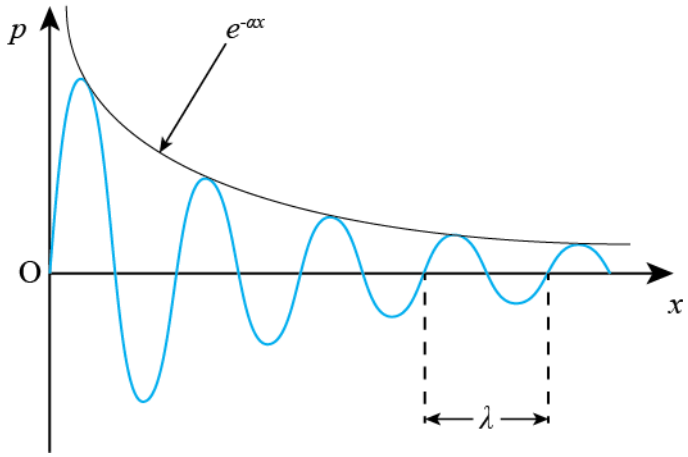

Figure 5 – Damped wave with wavelength λ and damping coefficient α.

The amplitude of a received acoustic signal is generally expressed in decibels relative to a reference pressure:

$$\mathrm{SPL} = 10\log_{10}\left(\frac{p}{p_{\mathrm{ref}}}\right)^2 = 20\log_{10} p - 20\log_{10} p_{\mathrm{ref}}, \qquad \textbf{Eq. 5}$$

where SPL is the sound pressure level in decibels. Decibels are always rounded to whole numbers. Table 1 gives some typical values for pressure changes and their respective level in decibels.

SPL [dB]	Multiplication
-20	0.10×
-12	0.25×
-6	0.50×
0	1×
6	2×
12	4×
20	10×
40	100×
60	1,000×
80	100,000×

Table 1 – Sound pressure levels and their corresponding multipliers.

Most acoustic waves propagate unhindered through the human body. A small proportion is specularly reflected on tissue transitions. The amount of reflected sound at such a boundary is dependent of the acoustic impedances on both sides of the boundary. The acoustic impedance *Z* of a medium is defined by

$$Z_i = \rho_i c_i, \qquad \textbf{Eq. 6}$$

where c_i and ρ_i are the speed and the density, respectively, of medium i. Reflection and transmission coefficients are used to predict reflections from boundaries.

In most organs, tissues have rather small acoustic impedance differences. The boundaries consist of cells with sizes much smaller than the wavelength of the ultrasound used for imaging. The signals travelling back to the sound source from tissue transitions are actually caused by scattering. Given the long wavelengths of the ultrasound, cells can be considered point scatterers. The backscattering from

point scatterers is proportional to the number of scatterers per volumetric unit (scattering density), proportional to the square of the combined compressibility and density differences of the scatterers, inversely proportional to the fourth power of the wavelength and therefore proportional to the fourth power of the frequency, and proportional to the sixth power of the radii of the point scatterers. For larger scatterers, such as collagens or veins, the scattering behaviour is different from the so-called Rayleigh scattering from point scatterers. The backscattering properties have been quantified for many structures of millimetre-size in organs. Using these quantifications of backscattered signal, abnormalities can be traced. As an example, fatty liver cirrhosis can be traced from the change in scattering from enlarged mean distances between lobular structures.

Moving scatterers such as blood cells create a shift in the ultrasound signal. This so-called Doppler shift can be approximated by

$$F_D \approx 2f\frac{v}{c}\cos\theta, \qquad \textbf{Eq. 7}$$

where θ is the angle between the ultrasound beam and the streaming direction (positive axis) and v is the magnitude of the streaming velocity (*cf*. Figure 6).

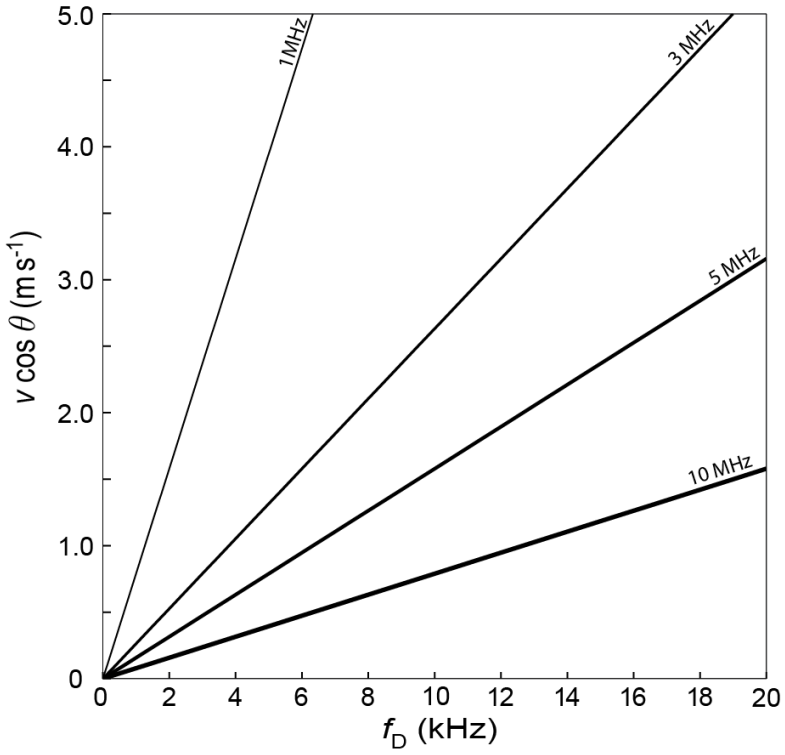

Figure 6 – Lateral velocity as a function of Doppler shift at four different transmitting frequencies.

Changes in the frequency of the signal are also caused by nonlinear propagation through tissue and by the presence of ultrasound contrast agents. Nonlinear propagation is caused by the fact that the speed of sound in compressed tissue is slightly higher than in extended tissue. Therefore, the peaks of ultrasound waves travel faster than the troughs. The waves are distorted farther away from the source, until only saw-tooth shapes remain. Figure 7 shows some waveforms at different distances from the source, and their frequency content.

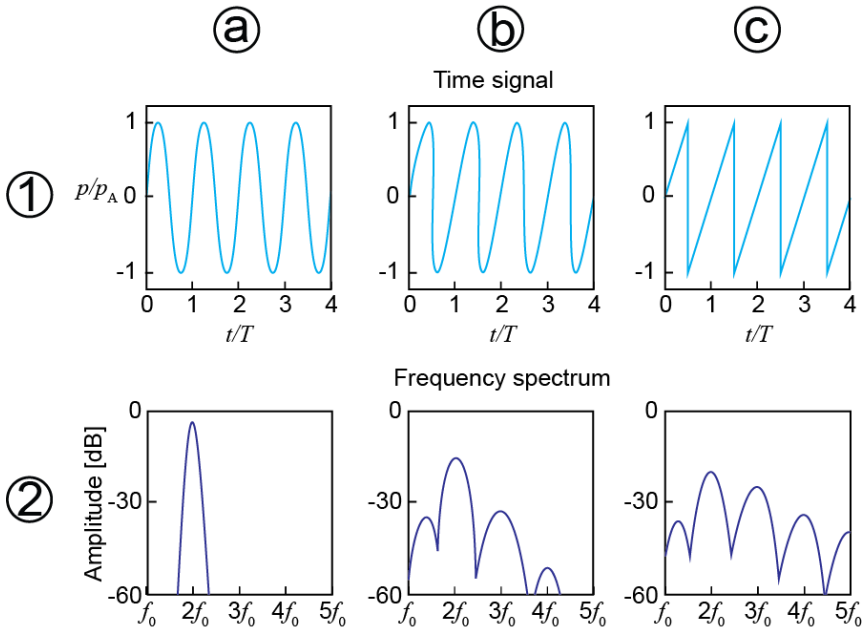

Figure 7 – Waveforms at different distances from the source, and their respective frequency spectra.

Blood cells are poor scatterers in diagnostic ultrasound. Because perfusion imaging is often desired in clinical diagnosis, ultrasound contrast agents have been injected to enhance the scattering from blood. Ultrasound contrast agents consist of microscopically small perfluorocarbon gas bubbles encapsulated by elastic (most commonly phospholipid) shells. These microbubbles oscillate linearly and nonlinearly in sound fields, radiating a detectable acoustic signal. Several detection strategies exist to reveal the presence of microbubbles and therefore blood (*cf*. Figure 8).

Recently, the peculiar behaviour of microbubbles under specific acoustic conditions close to living cells has led to research into therapeutic applications of microbubbles

whose shells have been modified to contain drugs or genes. Ultrasound-guided drug delivery might be possible using regular clinical ultrasound equipment.

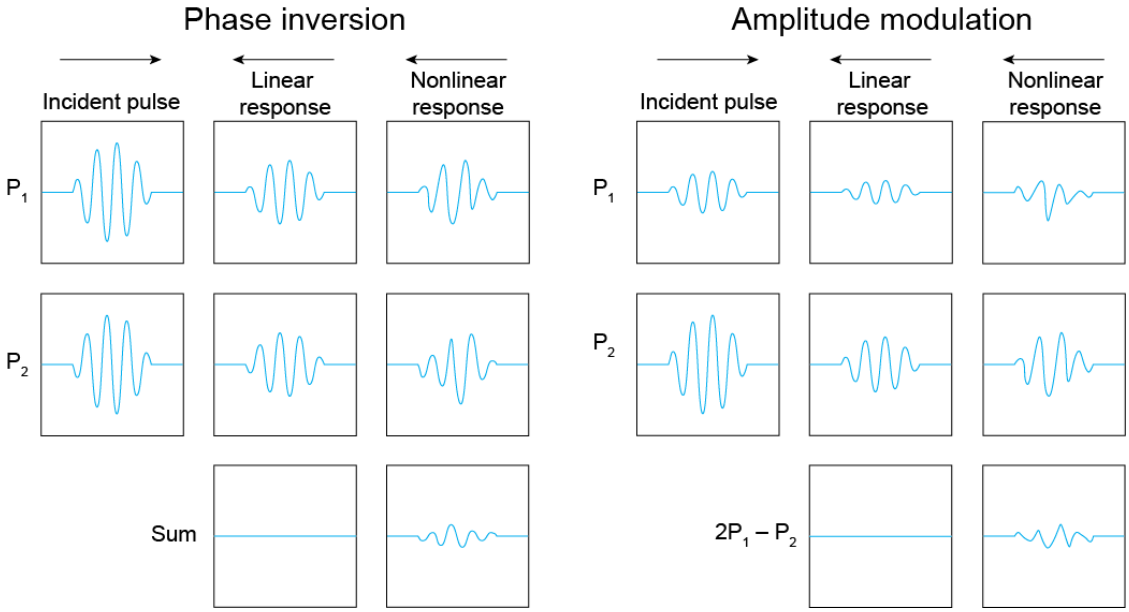

Figure 8 – Two detection strategies for the presence of microbubbles.

Transducers

Ultrasound transducers convert electrical signal to pressure waves and vice versa. With therapeutic devices, such as those used for physiotherapy or ultrasound-mediated surgery, only the transmit capability is used, whereas diagnostic devices both transmit and receive. In all cases, transducers contain piezoelectric elements to generate ultrasound or convert ultrasound into an electrical signal.

When strain is applied, the electric charges in the elements are redistributed, therefore generating an electrical impulse. Inversely, when an electrical impulse is applied, it changes the geometry of the piezoelectric material. This is true for all piezoelectric materials.

Apart from one or more piezoelectric elements with electrodes attached to both sides, ultrasound transducers consist of a backing behind the element, and one or more matching layers in front of the element (*cf.* Figure 9).

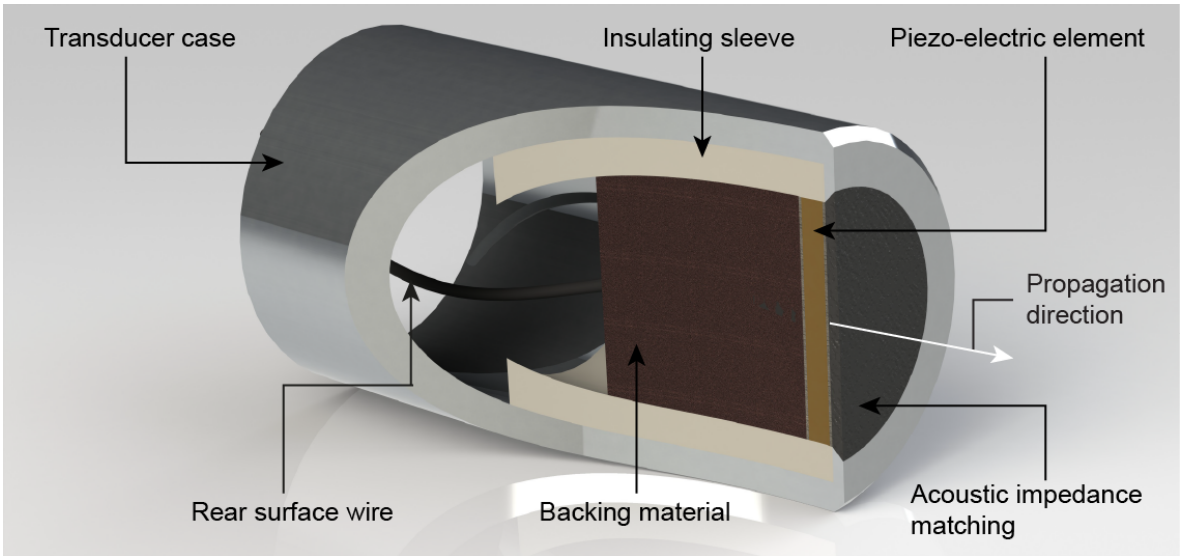

Figure 9 – Components of a single-element transducer.

The thickness of the element determines its resonance frequency: its natural oscillation frequency. Without backing present, the element can oscillate with maximum amplitude, *i.e.*, extend and contract, at this frequency. The backing material determines the bandwidth of the transducers. The bandwidth is the frequency band at which a transducer generates and receives sound (*cf.* Figure 10). The choice of backing material is critical for the performance of the transducer. The matching layer forms a near-lossless transition between the element and the medium.

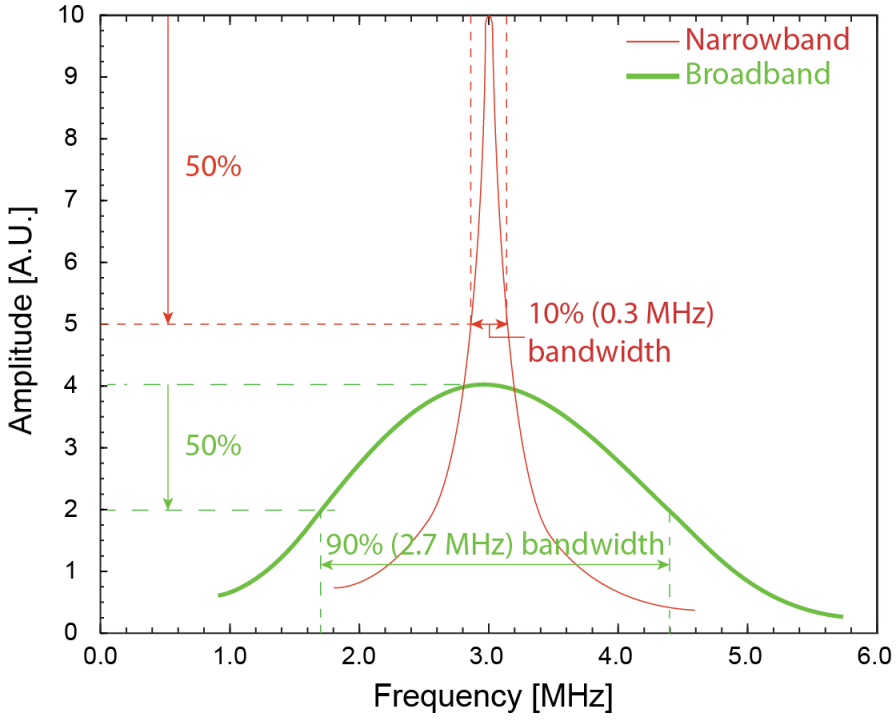

Figure 10 – Tradeoff between higher-power output and wide bandwidth.

A transmitting transducer creates a sound field. Close to the transducer surface, interference causes local pressure variations (*cf.* Figure 11). The width of this so-called near field is roughly equal to the diameter of the transducer. Its length is given by

$$N = \frac{D^2}{4\lambda}, \qquad \textbf{Eq. 8}$$

where D is the transducer diameter and N is the near-field length. In the far field, the sound field propagates with an opening angle 2γ:

$$\sin\gamma = 1.22\,\frac{\lambda}{D}. \qquad \textbf{Eq. 9}$$

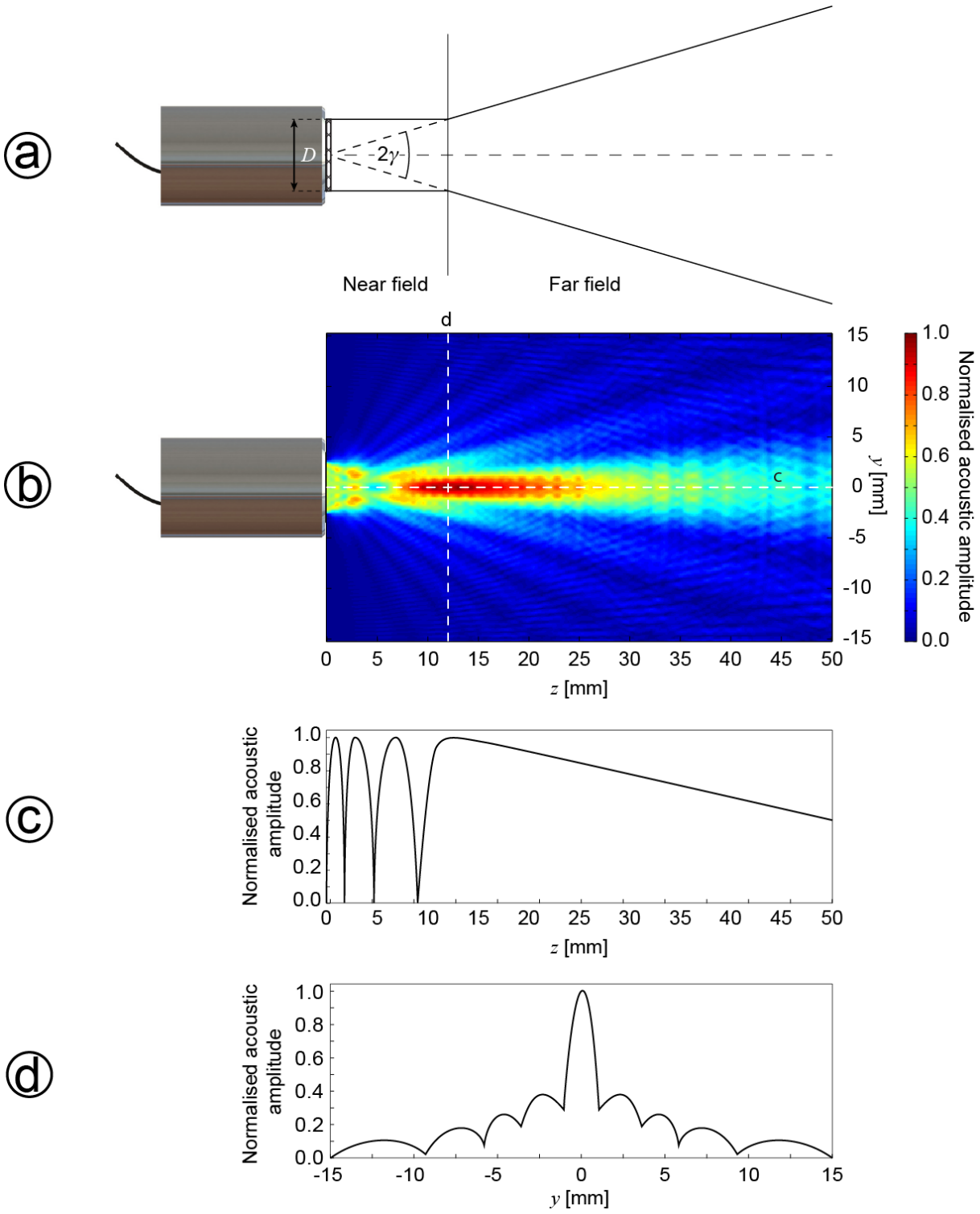

Figure 11 – Profiles of a sound field: (a) schematic; (b) 2D; (c) axial: (d) lateral.

The axial plane separating near field from far field is the natural focus of a transducer. A sound field can be further geometrically focussed by adding an acoustic lens to the transducer surface.

In clinical ultrasound machines, multi-element transducers are used (*cf.* **Figure 12**). These consist of arrays of transducer elements lined up in the lateral direction, which can be individually controlled, allowing for variable beam focussing (*cf.* **Figure 12**). Such so-called phased arrays exist in numerous layouts, including curvilinear, 1.5D, and 2D variations (*cf.* Figure 13).

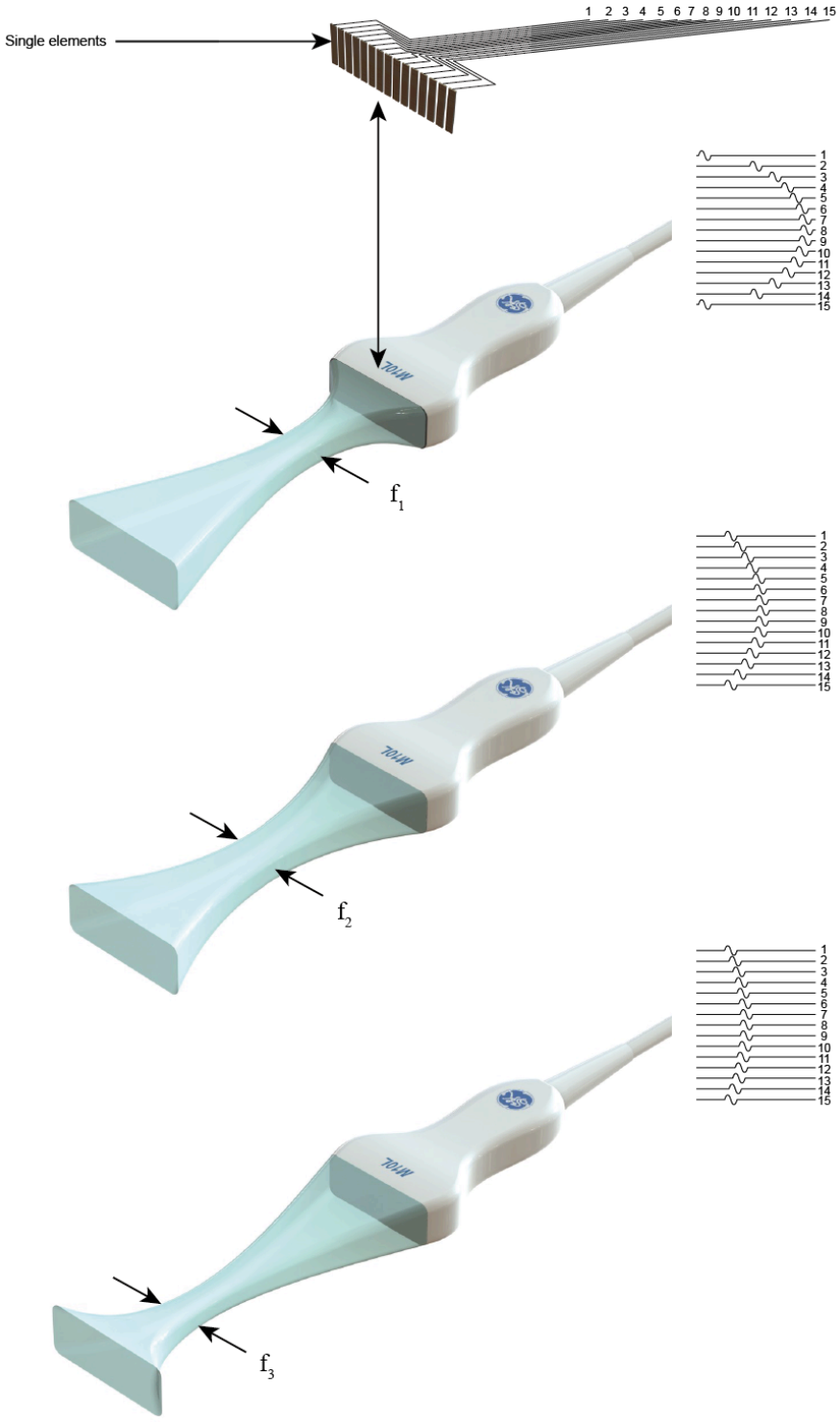

Figure 12 – Phased-array probes with variable focusing.

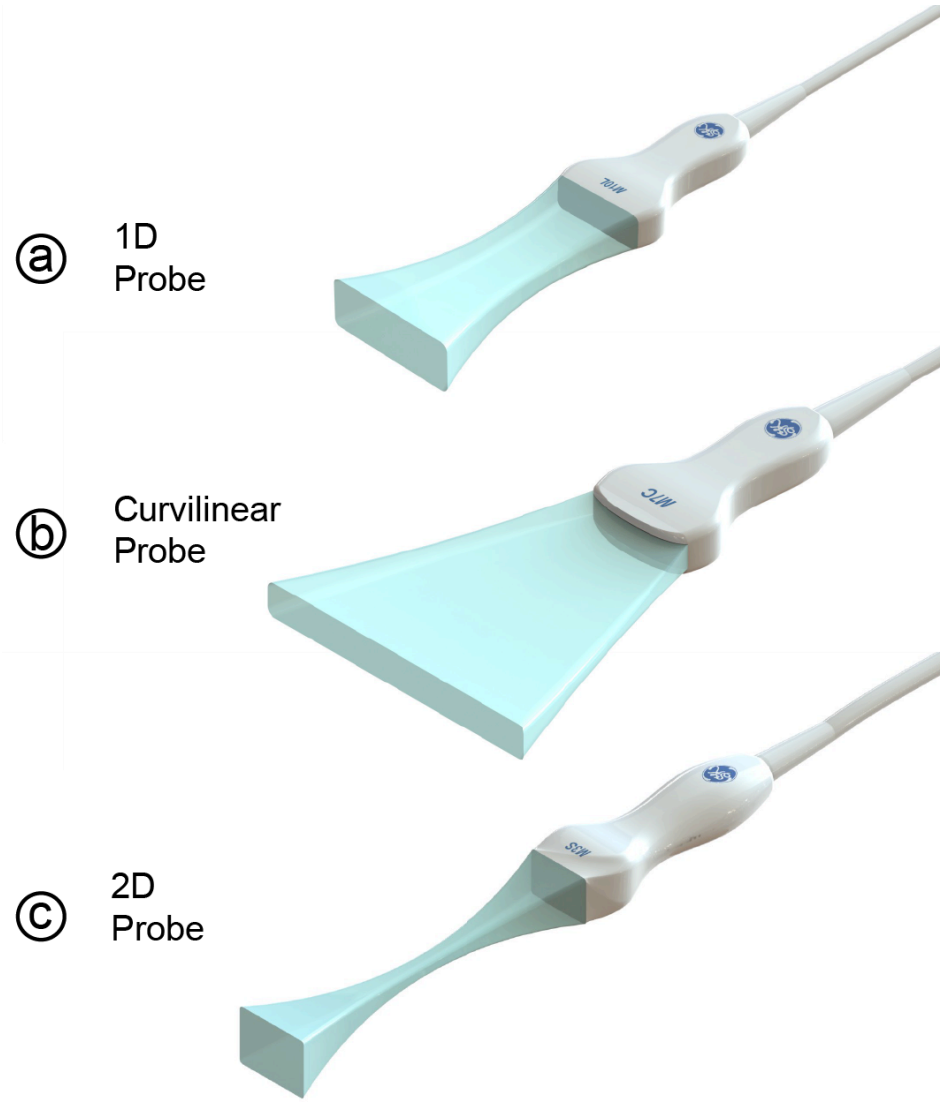

Figure 13 – Phased arrays and their beam profiles: (a) 1D; (b) curvilinear; (c) 2D.

Imaging

Unlike a continuous sound wave, ultrasound for diagnostic imaging is transmitted as a pulse sequence (*cf*. Figure 14). After transmission of a pulse with a certain centre frequency, backscattered signal from tissue is received until the next pulse is transmitted. The pulse repetition frequency (PRF) is the number of pulses per time unit. The duty cycle is the percentage of transmission time, equal to the pulse length times the PRF. The theoretical maximum distance of imaging is

$$R_{\max} = \frac{c}{2\,\mathrm{PRF}}. \qquad \textbf{Eq. 10}$$

Beware that the local speed of sound varies for different tissues. Therefore, ultrasound images built from the two-way travel times recorded are not converted to actual depth. The most commonly used mean speed of sound for quasi-depth conversion is 1540 m/s, which is the mean speed of sound in soft tissue (*cf*. Table 2). As this is just a chosen value, great care should be taken when drawing conclusions from quantitative spatial measurements using ultrasonic imaging.

Material/tissue	*c* [m/s]
Air	330
Silicon oil	980
Water	1490
Blood	1570
Fat	1460
Muscle	1580
Bone	3500
Mean in soft tissue	1540

Table 2 – Speed of sound for different biomaterials.

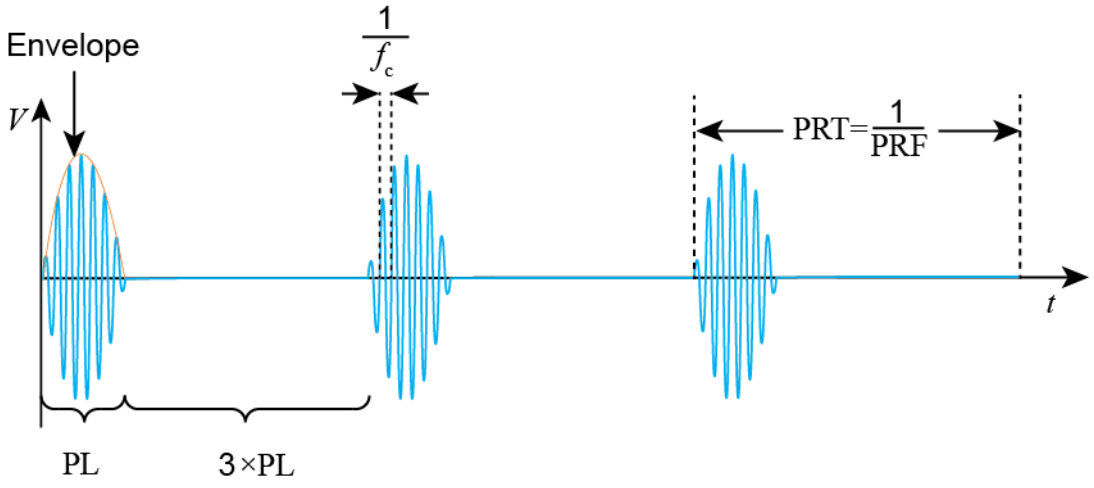

Figure 14 – Pulsed transmit signal, with a centre frequency f_c, pulse length PL, pulse repetition time PRT, pulse repetition frequency PRF. The duty cycle of this signal is 25%.

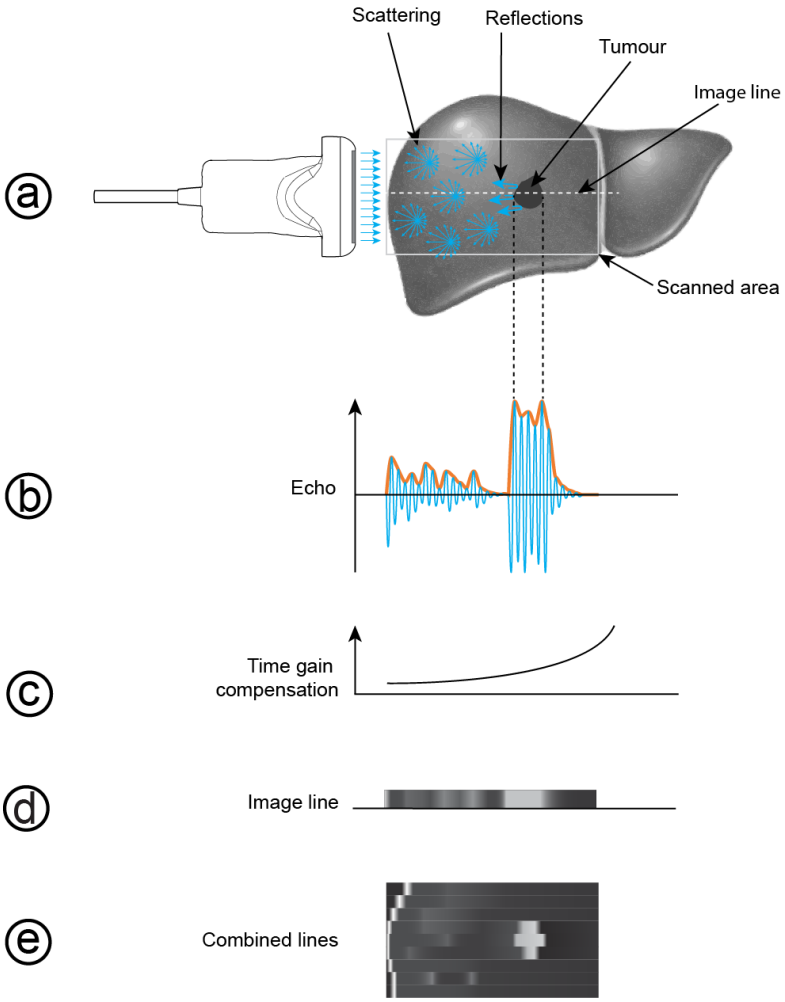

Figure 15 – Principle of B-mode imaging.

Figure 15 shows how a single beam is creating a line in an ultrasound image. If only the signal amplitude ("A") is considered, the imaging mode is called A-mode, whereas the amplitudes are represented by spots of brightness ("B") are more than one dimension, we speak of B-mode. Table 2 shows some typical applications of B-mode and the frequencies of choice. B-mode is the default clinical ultrasonic imaging method. B-mode is often combined with Doppler methods to track moving targets. Using probes with mechanically moving transducers or 2D arrays, 3D B-mode scans can be recorded.

Frequency [MHz]	Penetration depth [cm]	Target organ
2-3	30	Deep abdomen
4-5	20	Adult heart
12-15	3	Mammae, Thyroid, Endosonography
20-50	1	Eye

Table 3 – Some B-mode applications and their fundamental imaging frequencies.

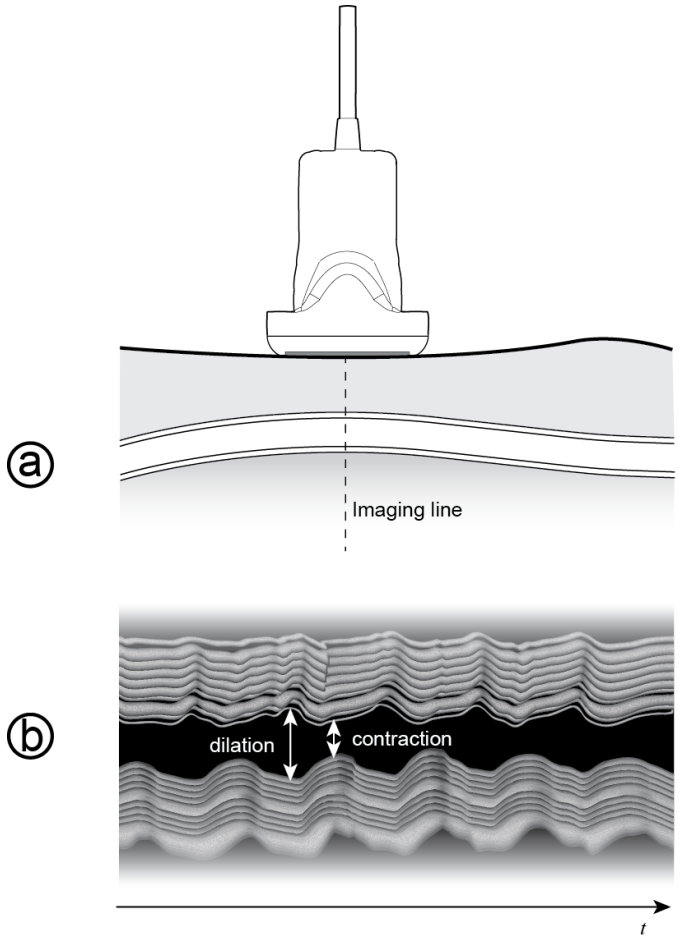

Figure 16 – M-mode imaging principle.

Displaying the received signal along one beam in motion ("M"), *i.e.*, as a function of time is called M-mode. The high time resolution allows for detailed study of periodically moving objects, such as the heart or a blood vessel (*cf*. Figure 16).

Resolution is the minimum distance between two points for them to be discriminated as separate points. The axial resolution is equal to the speed of sound divided through by twice the bandwidth. Hence, the wider the bandwidth (or the shorter the

pulse length) is, the smaller (better) the axial resolution is. The lateral resolution is proportional to the wavelength of the sound and the transducer focal depth, and inversely proportional to the transducer aperture.

Safety indices

The mechanical index (MI) gives an indication for the mechanical damage of tissue due to inertial cavitation: the ultrasound-induced formation of transient cavities:

$$\mathrm{MI} = \frac{\mathrm{PNP}}{\sqrt{f_c}}, \qquad \textbf{Eq. 11}$$

where PNP is the maximum value of the peak-negative pressure anywhere in the ultrasound field (measured in water but corrected for a different attenuation) normalised by 1 MPa and f_c is the centre transmit frequency normalised by 1 MHz.
At MI < 0.3, the acoustic amplitude is considered low enough for neonatal scans and pregnant women. At 0.3 < MI < 0.7, there is risk of minor damage to neonatal lung and intestine. At MI > 0.7, there is a theoretical risk of inertial cavitation and a more substantial risk if ultrasound contrast agents are being used.
Although the validity of the MI has been disputed, especially if an ultrasound contrast is used, there is currently no alternative available to judge the safety from cavitation-related damage in clinical settings.
Another, disputed, safety index is the thermal index (TI). It is a rough indicator of the temperature rise in tissue during ultrasound exposure, and defined by the ratio of the transmitted power and the estimated power needed to raise the tissue temperature 1°C. It should be noted that the TI does not indicate the actual temperature rise. Based on thermal indices, limitations to ultrasound exposure times have been recommended.
Near-future research will have to concentrate on redefining the safety standards.

Further reading

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Appendix

The equation of motion is

$$\frac{\partial p}{\partial x} = -\rho \frac{\partial^2 u}{\partial t^2}, \qquad \textbf{Eq. A 1}$$

where ρ is the density of the medium.

The linear 1-dimensional wave equation gives the sound pressure as a function of space and time:

$$\frac{\partial^2 p}{\partial t^2} = \frac{\kappa}{\rho}\frac{\partial^2 p}{\partial x^2} = c^2 \frac{\partial^2 p}{\partial x^2}, \qquad \textbf{Eq. A 2}$$

The speed of sound $c = \lambda f = \sqrt{\frac{\kappa}{\rho}}$ is a material property of the medium. Here, κ is the bulk (incompressibility) modulus. Solutions of the wave equation have the form:

$$p(x,t) = p_{\mathrm{A}} \sin 2\pi(ft - kx) = p_{\mathrm{A}} \sin 2\pi\left(\frac{t}{T} - \frac{x}{\lambda}\right), \qquad \textbf{Eq. A 3}$$

where $k = \frac{2\pi}{\lambda}$ is the wave number and p_{A} is the pressure amplitude.