Medical diagnostic ultrasound

By Jens E. Wilhjelm and Ole Trier Andersen

Ørsted•DTU, Ørsteds plads, building 348

Technical University of Denmark

2800 Kgs. Lyngby

1 Introduction

Medical diagnostic ultrasound is an imaging modality that makes so-called tomographic images (tomo = *Gr. tome, to cut* and graphic = *Gr. graphein*, to write). It is a diagnostic modality, *i.e.* it concerns gathering of information without modifying, in any way[1], the biological medium from which it gathers information. Ultrasound is sound with a frequency over the audible range (20 Hz - 20 kHz), and the frequencies normally applied in clinical imaging are between 1 MHz and 20 MHz. The sound is generated by a transducer that first acts as a loudspeaker sending out an acoustic pulse in a given direction and subsequently acts as a microphone in order to record the acoustic echoes generated along the path of the emitted pulse. These echoes carry information about the acoustic properties of the tissue along the path. The emission of acoustic energy and the recording of the echoes normally take place at one and the same transducer, in contrast to CT imaging, where the emitter and recorder are located on each side of the patient.

This chapter will explain the generation and reception of ultrasound and the different types of waves. This is followed by a description of ultrasound's interaction with the medium, which gives rise to the echo information that is used to make images. The different kinds of imaging modalities is next described, finalized with a description of spatial compounding. The chapter is concluded with a list of symbols, terms and references.

2 Generation of ultrasound

Ultrasound (as well as sound) needs a medium, in which it can propagate by means of local deformation of the medium. One can think of the medium as being made of small spheres (*e.g.* the molecules in some cases), that are connected with springs. When mechanical energy is transmitted through such a medium, the spheres will oscillate around their resting position. Thus, the propagation of sound is due to a continuous interchange between kinetic energy and potential energy, related to the density and the elastic properties of the medium, respectively.

The two most simple waves that can exist in solids are longitudinal waves in which the particle movements occur in the same direction as the propagation (or energy flow), and transversal (or shear waves) in which the movements occur in a plane perpendicular to the propagation direction. In water

1. To obtain acoustical contact between the transducer and the skin, a small pressure must be applied from the transducer to the skin. In addition to that, ultrasound scanning causes a very small heating of tissue (less than 1°C)

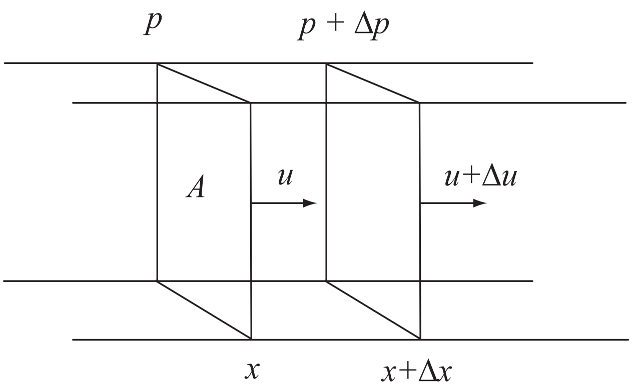

Figure 1 1D situation showing a liquid element inside a sound wave.

and soft tissue the waves are mainly longitudinal. The frequency, f, of the particle oscillation is related to the wavelength, λ, and the propagation velocity c:

$$\lambda f = c \tag{1}$$

The sound speed in soft tissue at 37°C is around 1540 m/s, thus at a frequency of 7.5 MHz, the wavelength is 0.2 mm.

An acoustic wave is normally characterized by its pressure. Thus, in order to obtain a quantitative relation between the particle velocity in the medium, u, and the acoustic pressure, p, a simple situation with 1D propagation in a loss-less media will be considered, as shown in Figure 1. This figure shows a volume element of length Δx and with cross-sectional area A. The volume will be $V = A\Delta x$. The density of the medium - a liquid, for instance, - is ρ and the mass of the element will then be $\rho A \Delta x$.

The pressure p is a function of both x and t. Consider the variation in space first: There will be a pressure difference, Δp, from x to $x{+}\Delta x$, thus the volume element will be subject to a force $-A\Delta p$. By applying Newton's second law ($F = ma$):

$$-A\Delta p = \rho A \Delta x \frac{du}{dt} \tag{2}$$

or after performing the limit ($\Delta \rightarrow$ d)

$$\frac{dp}{dx} = -\rho \frac{du}{dt} \tag{3}$$

Next consider the variations over a time interval Δt. A difference in velocity, Δu, between the front surface (at x) and the back surface (at $x{+}\Delta x$) of the elemental volume will result in a change in that volume which is:

$$\Delta V = A((u + \Delta u)\Delta t - u\Delta t) = A \Delta u \Delta t \tag{4}$$

which in turn is connected with a change in pressure, Δp, according to

$$\Delta V = -\kappa (A \Delta x) \Delta p \tag{5}$$

where κ is the compressibility of the material (*e.g.* a liquid) in units of Pa^{-1}. Performing the same limit as above, gives the second equation:

$$\frac{du}{dx} = -\kappa \frac{dp}{dt} \tag{6}$$

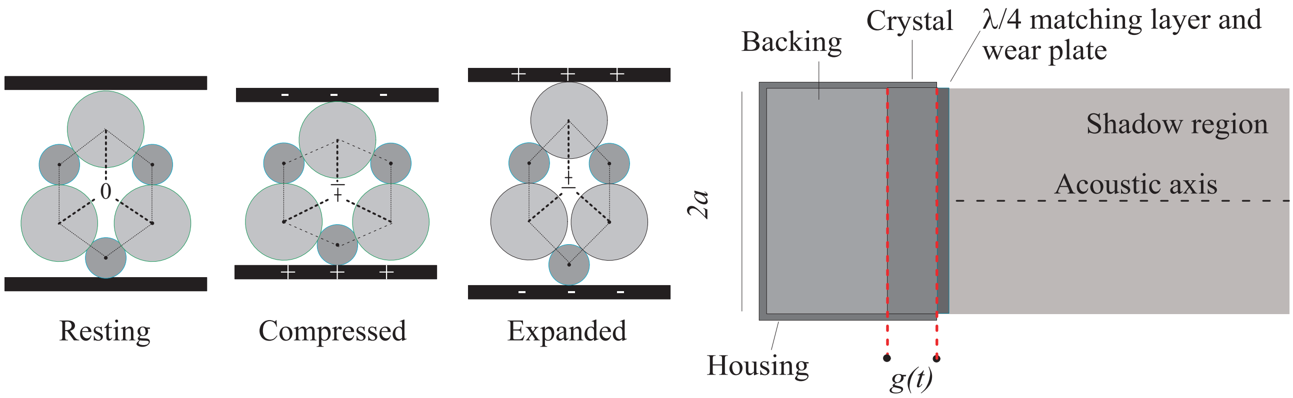

Figure 2 Left: Piezo electric crystal at different states. Right: Single element transducer consisting of piezoelectric crystal with electrodes. This "sandwich" is placed between a backing material and the matching layer towards the medium.

Equations (3) and (6) are the simplest form of the wave equations describing the relation between pressure and particle velocity in a loss-less isotropic medium.

2.1 Piezoelectricity

The acoustic field is generated by using the piezo electric effect present in certain ceramic materials. If electrodes (*e.g.* thin layers of silver) are placed on both sides of a disk of such a material, and the material is compressed or expanded, the displacement of charge inside the material will cause a voltage change on the electrodes, as illustrated in Figure 2. Likewise, a forced change in voltage across the two electrodes, will result in a physical deformation of the crystal. This is used for reception and emission of acoustic energy, respectively.

2.2 The acoustic field from a disk transducer

It is assumed that the piezoelectric, disk-shaped crystal is fixed at the back, as illustrated in Figure 2 (right) and can move freely at the front. Specifically, movement of the surface of the transducer can be described by a velocity vector oriented perpendicular to the surface. In short, the electrical signal applied to the transducer is converted by the transfer function of the transducer to a velocity function describing the movement of the transducer surface. Note the backing material located behind the crystal; this is used to dampen the free oscillation of the crystal, thereby creating a short vibration, when an impulse is applied to the crystal. The radius of the crystal is denoted a. The thickness of the crystal is selected according to the frequency of operation so that it is $\lambda_{piezo}/2$, where λ_{piezo} is the wavelength of sound in the crystal material.

In order to assess the pressure field generated by the transducer, one can assume that the surface of the crystal can be divided up into many small surface elements, each contributing to the entire pressure field. If the contribution to the final pressure at a given field point can be found for a single surface element, then the total pressure field can be calculated by using Huygens'[1] principle to sum up all contributions. Rather than doing this analytically, a graphical illustration is provided in Figure 3 which shows point sources along a diameter of the transducer disk. For each point source, a bow shows the location of the equal-phase-fronts of the spherical pressure wave generated from that source at given instances in time.

1. Christian Huygens, physicist from the Netherlands, 1629-95.

The wave fronts generated by the flat transducer in Figure 3 (left) tend towards a (locally) plane wave inside the shadow of the transducer. The pressure field is thus broad, and unsuitable for imaging purposes, as will become clear, when the imaging technique is considered later. In order to focus the ultrasound field and obtain a situation where the acoustic energy travels along a narrow path, a focused transducer is used, as illustrated in Figure 3 (right). In this situation, the individual spherical waves from the transducer are performing constructive interference at the focal point, whereas at all other points, the interference is more or less destructive. In order to make this work efficiently, the wavelength must be much smaller than the distance to the focal point. However, a typical depth of the focal point for a 7.5 MHz transducer - 20 mm - will correspond to 100λ.

The interference phenomena can be explored in everyday life: if one positions oneself with one ear pointing into a loudspeaker and turns up the treble, then the sound picture will change while you move back and forth. What happens is that the ear is moved to different points in space, which exhibits different amounts of constructive and destructive interference. This phenomenon is less distinct at low frequencies, because the wavelength gets larger.

As noted above, dimensions give most insight, when they are measured in wavelength. Consider the planar transducer in Figure 3 (left): The near field from this type of transducer is defined as[4] the region between the transducer and up to a range of a^2/λ. The far field region corresponds to field points at ranges *much larger* than a^2/λ. In Figure 3 (left), a is specified, but λ is not. If the transducer frequency is f_0 = 0.5 MHz, then a^2/λ = 33 mm, which is in the middle of the plot. If f_0 = 7.5 MHz, then a^2/λ = 0.5 m! The explanation is as follows: The far field is defined as the region, where there is only moderate to little destructive interference. If this should be possible, then from a given field point in this region, the distance to any point on the transducer surface should *vary* much less than a wavelength. Specifically, consider a given field point not on the acoustic axis. Next, draw two lines to the two opposite edges of the transducer. Now the difference in length of these two lines - measured in wavelength - must be much less than one, in order to have little destructive interference at this field point. Thus, the higher the frequency, the lower the wavelength, and the farther away one must move from the transducer surface in order to get differences between the length of the two lines much less than one wavelength.

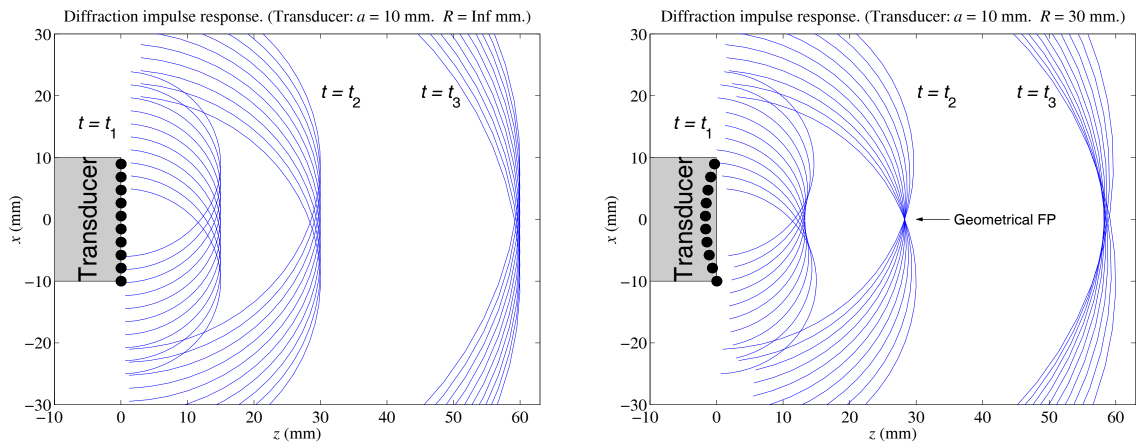

Figure 3 Left: Circles showing wave fronts of equal phase at three different time instances from part of the transducer split up into point sources. For simplicity, only the point sources that are located on the vertical diameter are shown. Right: The same for a focused transducer. FP = focal point.

3 Types of ultrasound waves

An ultrasound field from a physical transducer will always show a complicated behaviour as can be sensed from Figure 3. Each point source emits exactly the same pressure wave. The temporal form of this pressure wave could be as exemplified in Figure 6. Thus, the circles in Figure 3 indicate examples of spatial and temporal locations of each of the individual waveforms that have to be added in order to construct the total pressure field in front of the transducer (however, the circles in Figure 3 only represent point sources on a single diameter across the transducer; many more point sources would be needed to represent the total field from a disk transducer).

In order to obtain some tools to better describe and understand these fields, wave theory makes use of two types of simple, yet theoretical, waves which are introduced here:

The *plane wave*, where any field parameter is constant in a plane perpendicular to the propagation direction. As a plane extent over the entire space, it is not physically realizable (but within a given space, an approximation to a plane wave can be obtained locally, such as in the shadow of a planar transducer). The other type is a *spherical wave*. It originates from a point (source) and all acoustic parameters are constant at spheres centered around this point.

As a further restriction of the plane wave field, it could be considered monochromatic, that is, it oscillate at a single frequency, f_0. The equation for such a wave in 1D is:

$$p(x,t) = P_0 \exp(-\mathrm{j}\,(2\pi f_0 t - 2\pi x/\lambda)) \tag{7}$$

where P_0 is the pressure magnitude (units in pascal, Pa) x is the distance along the propagation direction and $\lambda = c/f_0$. (7) is a complex sinusoidal that depends on space and time.

An important concept in wave theory is *diffraction*. Ironically, the term diffraction can best be described by what it is *not*: “Any propagating scalar field which experiences a deviation from a rectilinear propagation path, when such deviation is not due to reflection or refraction, is generally said to undergo diffraction effects. This description includes the bending of waves around objects in their path. This bending is brought about by the redistribution of energy within the wave front as it passes by an opaque body.”[3] Examples where diffraction effects are significant are: Propagation of waves through an aperture in a baffle (*i.e.* a hole in a plate) and radiation from sources of finite size.[3] With the above definition, the only non-diffracted wave is the plane wave.

4 Ultrasound’s interaction with the medium

4.1 Reflection and transmission

The plane wave can be used when explaining reflection and transmission from one medium to another, when the interface between the media is planar. For this purpose the *specific acoustic impedance*, z, is introduced. In a homogeneous medium it is defined as the ratio of pressure to particle velocity in a progressing plane wave, and can be shown to be the product of the density, ρ, and acoustic propagation velocity c of the medium. Thus, medium 1 is specified in terms of its density, ρ_1, and acoustic propagation velocity c_1. The specific acoustic impedance for this medium is $z_1 = \rho_1 c_1$, with units: kg/(m^{2}s). Likewise for medium 2: $z_2 = \rho_2 c_2$. The interaction of ultrasound with this interface is illustrated in Figure 4, where an incident plane wave is reflected and transmitted at the interface between medium 1 and medium 2. The (pressure) reflection coefficient between the two media is:

$$R = \frac{z_2/(\cos\theta_t) - z_1/(\cos\theta_i)}{z_2/(\cos\theta_t) + z_1/(\cos\theta_i)} \tag{8}$$

where the angle of incidence, θ_i, and transmission, θ_t, are related to the propagation velocities as

$$\frac{\sin\theta_i}{\sin\theta_t} = \frac{c_1}{c_2}. \tag{9}$$

Equation (9) is a statement of Snell's law,[2] which also states that: $\theta_r = \theta_i$. The pressure transmission coefficient is $T = 1 + R$.

It should be noted here, that Snell's law in optics are valid for rays, while it is valid for plane waves in acoustics. However, when the acoustic wave travels like a beam, Snell's law is approximately valid. The validity is related to the properties of the beam, namely to which degree the wave field inside the beam can be considered locally plane (which again is related to the thickness of the beam, measured in wavelengths).

Strictly speaking, if the field incident on an interface is not fully planar, and the interaction is to be modelled, then the field should be decomposed into a number of plane waves, just like a temporal pulse can be decomposed into a number of infinite tone signals. The plane waves can then be reflected one by one, using (8) and (9).

Biologically, an example of approximate reflection can be observed at the interface between blood and the intima of vessel walls.

4.2 Scattering

While reflection takes place at interfaces of infinite size, scattering takes place at small objects with dimensions much smaller than the wavelength. Here, the specific acoustic impedance of the small object is different from the surrounding medium. The scattered wave will be more or less spherical, and thus propagate in all directions, including direction towards the transducer. The latter is denoted *back*-scattering.

The scattering from particles much less than a wavelength is normally referred to as Rayleigh scattering. The intensity of the scattered wave increases with frequency to the power of four.

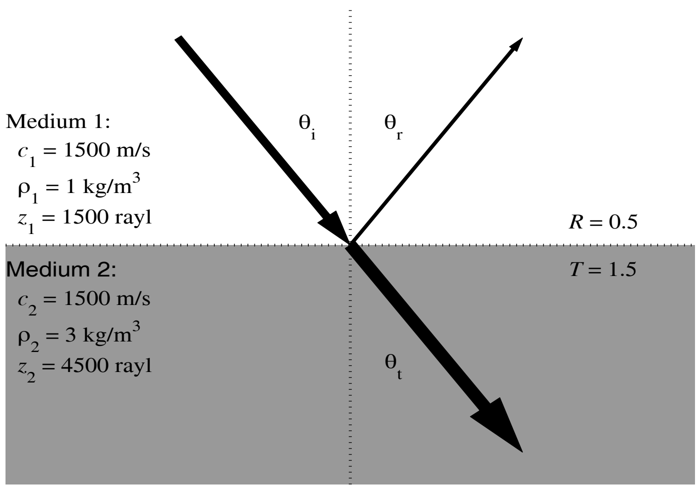

Figure 4 Graphical illustration of Snell's law describing the incident plane wave, reflected plane wave and transmitted (refracted) plane wave. The three arrows indicates the propagation *direction* of the plane waves. The thickness of an arrow is proportional to the magnitude of the pressure of the wave it represents.

Biologically, scattering can be observed in most tissue and especially blood, where the red blood cells are the predominant cells. They have a diameter of about 7 μm much smaller than the wavelength of clinical ultrasound.

4.3 Absorption

Absorption is the conversion of acoustic energy into heat. The mechanisms of absorption are not fully understood, but relate, among other things, to the friction loss in the springs, mentioned in Subsection 2. More details on this can be found in the literature.[2]

Absorption by itself can be observed by sending ultrasound through a viscous liquid such as oil.

4.4 Attenuation

The loss of intensity (or energy) of the forward propagating wave due to reflection, refraction, scattering and absorption is denoted attenuation. The intensity is a measure of the power through a given cross section; thus the units are W/m^2. It can be calculated as the product between particle velocity and pressure: $I = pu = p^2/z$. If $I(0)$ is the intensity of the pressure wave at some reference point in space and $I(x)$ is the intensity at a point x further along the propagation direction then the attenuation of the acoustic pressure wave can be written as:

$$I(x) = I(0)e^{-\alpha x} \tag{10}$$

where α (in units of m^{-1}) is the attenuation coefficient. α depends on the tissue type (and for some tissue types like muscle, also on the orientation of the tissue fibres) and is approximately proportional with frequency.

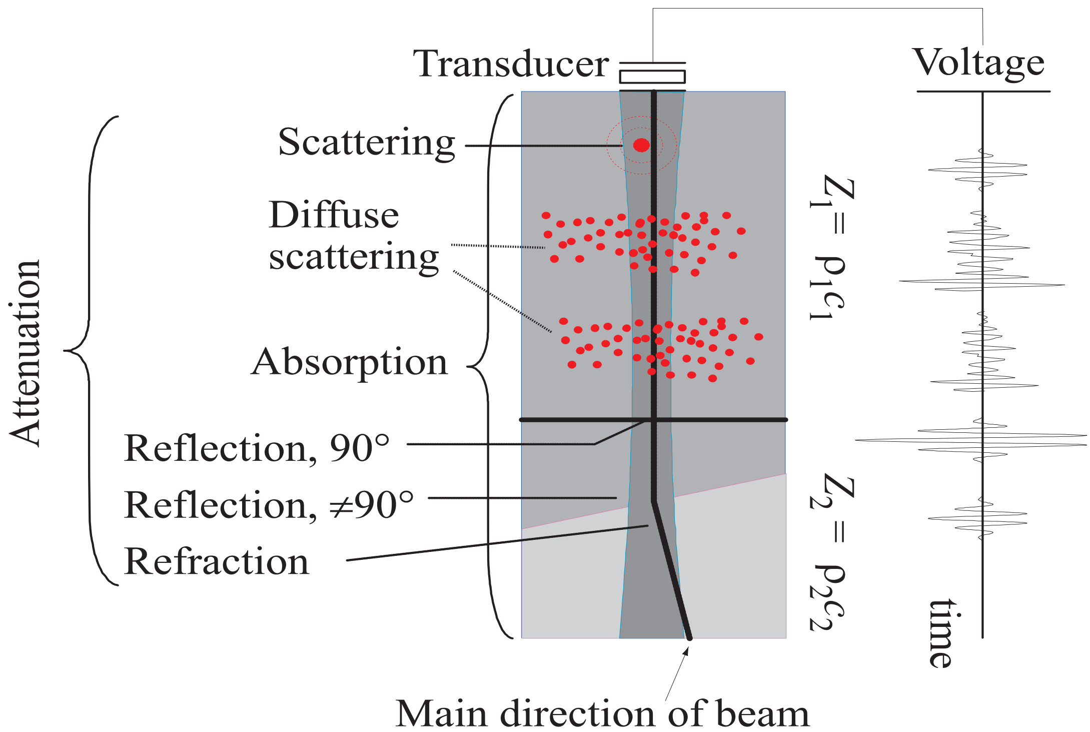

Figure 5 Interaction of ultrasound with tissue. The left drawing shows the medium with the transducer on top. The ultrasound beam is shown superimposed onto the medium. The right part of the drawing shows the corresponding received echo signal.

As a rule of thumb, the attenuation in biological media is 1 dB/cm/MHz. As an example, consider ultrasound at 7.5 MHz. When a wave at this frequency has travelled 5 cm in tissue, the attenuation will (on average) be 1 dB/cm/MHz x 5 cm x 7.5 MHz = 37.5 dB. For bone, the attenuation is about 30 dB/MHz/cm. If these two attenuation figures are converted to intensity half-length (the distance corresponding to a loss of 50 %) at 2 MHz, it would correspond to 15 mm in soft tissue and 0.5 mm in bone.

4.5 An example of ultrasound's interaction with biological tissue

When an ultrasound wave travels in a biological medium all the above mechanisms will take place. Reflection and scattering will not take place as two perfectly distinct phenomena, as they were described above. The reason is that the body does not contain completely smooth interfaces of infinite size. Likewise, the scattered wave from infinitesimally small point objects will also be infinitesimally small in amplitude and thereby not measurable.

The scattered wave moving towards the transducer as well as the reflected wave moving towards the transducer will be denoted *the echo* in this book.

The effects in Subsection 4.1 - 4.4 are illustrated in Figure 5.

The absorption continuously takes place along the acoustic beam, as media 1 and media 2 (indicated by their specific acoustic impedances) are considered lossy.

Consider the different components of the medium: Scattering from a single inhomogeniety is illustrated at the top of the medium. Below is a more realistic situation where the echoes from many scatterers create an interference signal. If a second identical scattering structure is located below the first, then the interference signal will be roughly identical to the interference signal from the first. The overall amplitude, however, will be a little lower, due to the absorption and the loss due to the first group of scatterers. Notice that the interference signal varies quite a bit in amplitude.

The emitted signal next encounters a thin planar structure, resulting in a well-defined strong echo.

Next, an angled interface is encountered, giving oblique incidence and thus refraction, according to (8) and Figure 4. The change in specific acoustic impedance is the same as above, but due to the non-perpendicular incidence, less energy is reflected back. The transmitted wave undergoes refraction, and thus scatterers located below this interface will be imaged geometrically incorrectly.

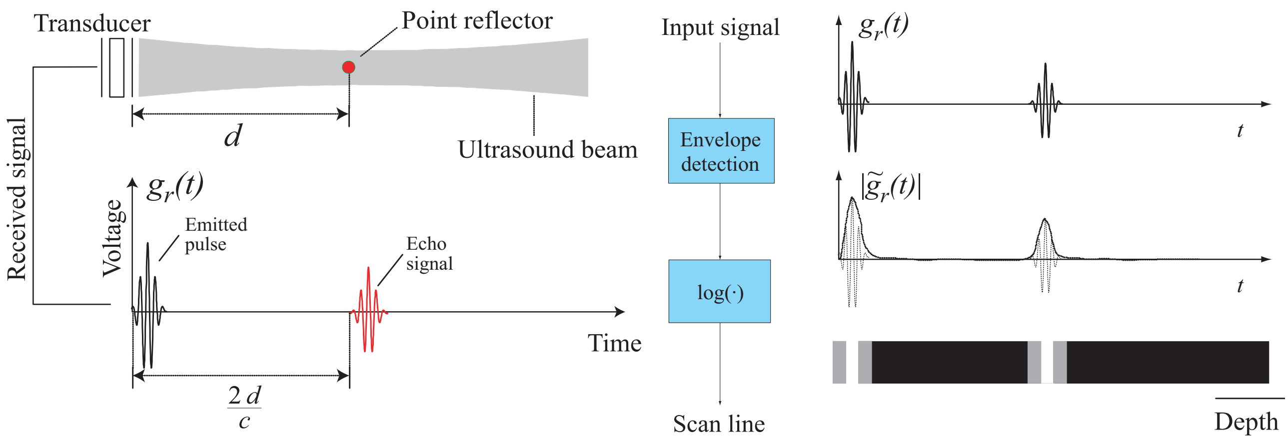

Figure 6 The basic principle behind pulse-echo imaging. An acoustic pulse is emitted from the transducer, scattered by the point reflector and received after a time interval, equal to the round trip travel time.

5 Imaging

5.1 A-mode

The basic concept behind medical diagnostic ultrasound is shown in Figure 6, which also shows the simplest mode of operation, A-mode. In the situation in Figure 6 (left) a single point scatterer is located in front of the transducer at depth d. A short pulse is emitted from the transducer, and at time $2d/c$, the echo from the point target is received by the same transducer. Thus, the deeper the point scatterer is positioned, the later the echo from this point scatterer arrives. If many point scatterers (and reflectors) are located in front of the transducer, the total echo can be found by simple superposition, as this is a linear system, when the pressure amplitude is sufficiently low.

The received signal, $g_r(t)$, is Hilbert transformed to $g_{rH}(t)$ in order to create the corresponding analytical signal $\tilde{g}_r(t) = g_r(t) + jg_{rH}(t)$. Twenty time the log of the envelope of this signal, $20\log|\tilde{g}(t)|$, is then the envelope in dB, which can be displayed as a gray scale line, as shown in Figure 6 (right). Such a gray scale bar is called a scan line, which is also the word used for the imaginary line in tissue, along which $g_r(t)$ is recorded. Note, that because the envelope process is not fully linear, the scanner does not constitute a fully linear system.

5.2 M-mode

If the sequence of pulse emission and reception is repeated infinitely, and the scan lines are placed next to each other (with new ones to the right), motion mode, or M-mode, is obtained. The vertical axis will be depth in meters downwards, while the horizontal axis will be time in seconds pointing to the right. This mode can be useful when imaging heart valves, because the movement of the valves will make distinct patterns in the “image”.

5.3 B-mode

Brightness or B-mode is obtained by physically moving the scan line to a number of adjacent locations. The principle is shown in Figure 7. In this figure, the transducer is moved in steps mechanically across the medium to be imaged. Typically 100 to 300 steps are used, with a spacing between 0.25λ and 5λ. At each step, a short pulse is emitted followed by a period of passive registration of the echo. In order to prevent mixing the echoes from different scan lines, the registration period has to be long enough to allow all echoes from a given emitted pulse to be received. This will now be considered in detail.

Assume that the average attenuation of ultrasound in human soft tissue is a in units dB/MHz/cm. If the smallest echo that can be detected - on average - has a level of γ in dB, relative to the echo from tissue directly under the transducer, then the maximal depth from where an echo can be expected is $\gamma = a f_0 2D_{max}$ or

$$D_{max} = \frac{\Upsilon}{2af_0} \tag{11}$$

Example: According to a rule of thumb, the average attenuation of ultrasound in human soft tissue is 1 dB/MHz/cm. Assume that $\gamma = 80$ dB. At $f_0 = 7.5$ MHz (11) gives $D_{max} = 5.3$ cm.

The time between two emissions will then be $T_r = 2D_{max}/c$, which is the time it take the emitted pulse to travel to D_{max} and back again. If there are N_l scan lines per image, then the frame-rate (number of images per second produced by the scanner) will be

$$f_r = (T_r N_l)^{-1}. \tag{12}$$

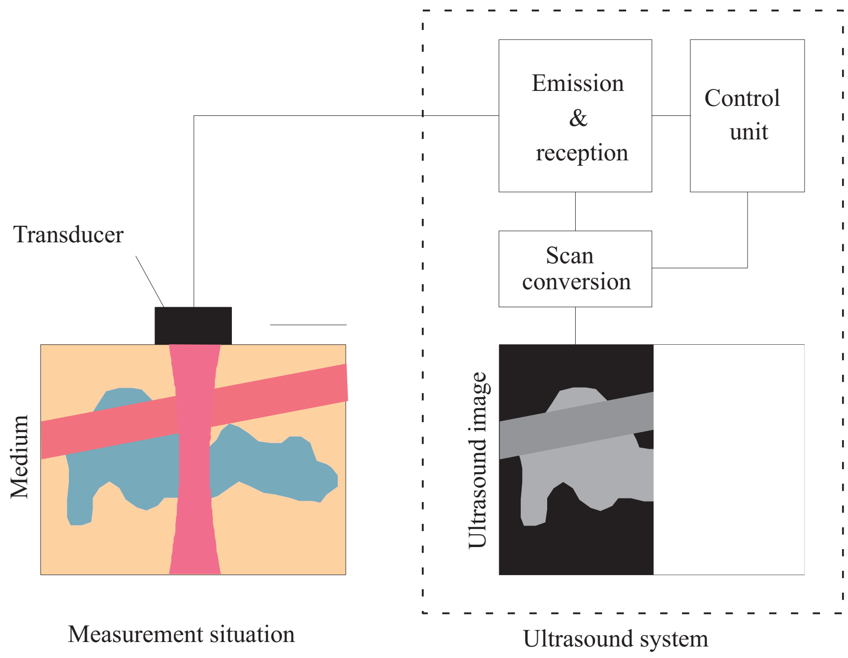

Figure 7 The principle of a simple B-mode ultrasound system. At this particular point in time, half of the image has been recorded.

Example: For $N_l = 200$, $f_r = 70$ Hz a good deal more than needed to obtain "real-time" images (some 20 frames per second). However, an f_r of 70 Hz might not be more than an adequate temporal resolution, when studying heart valves. If the total image width is 40 mm, then the distance between adjacent scan lines is 40 mm / 200 = 0.2 mm. Please note that this number is not directly reflecting the spatial resolution size of the scanner, which can be studied in Chapter 7.

6 Array transducers

The recording of a B-mode ultrasound image by mechanical movement of the transducer is now an old technique. Today most ultrasound systems apply array transducers, which consist of up to several hundreds of crystals, arranged along a straight or curved line. Each element of the transducer array, or a subset of elements, are connected to a complete transmitter/receiver, giving from 32 to several hundred independent channels. The shape, direction and location of the ultrasound beam can then be controlled electronically (in the newest scanners completely by software) thereby completely eliminating mechanical components of the transducer. In the most flexible systems, the amplitude, waveform and delay of the pulses can be controlled individually and precisely.

Two different types of transducer systems exist: *Phase array systems*, where all elements are in use all the time. The beam is then steered in different directions to cover the image plane. In the *linear array systems*, a subset of elements are used. From this subset a beam is created, and then translated by letting the subset of elements "slide" over the entire array.

7 Resolution size and point spread function

The resolution size of an imaging system can be assessed in many different ways. One way is to record an image of a small point target. The resulting image will then be the *point spread function* (psf), *i.e.* an image which shows how much a point target is "spread out", due to the imaging system. The point target should be much smaller than the smallest size of the psf. Another related way is to

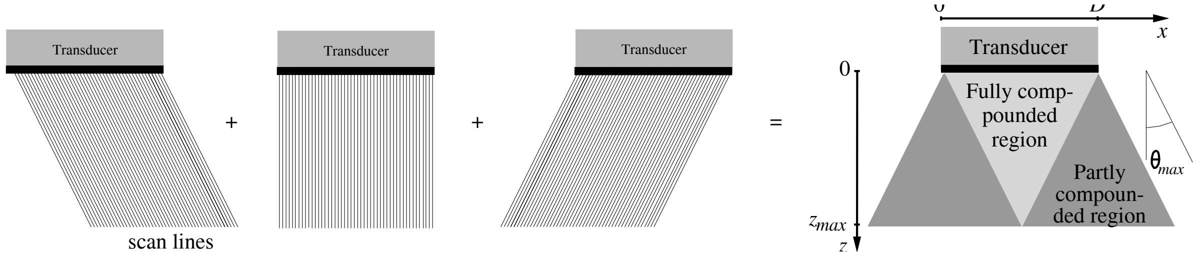

Figure 8 The principle of spatial compound imaging for $N_\theta = 3$. Three single-angle images are recorded from three different angles and then averaged to form the compound image. Inside the triangular region, the image is fully compounded, outside, less compounded.

image two point targets with different separations, and see how close they can be positioned and still be distinguishable.

The –3 dB width of the psf in the vertical and horizontal image direction will then be a quantitative measure for the resolution size. The two directions corresponds to the depth and lateral direction in the recording situation, respectively.

The resolution in the depth direction (axial resolution) can be appreciated from the echo signal in Figure 6. This signal was created by emitting the smallest number of periods. Because axial resolution can be improved only by decreasing the length of the echo signal from the point target, the centre frequency of the transducer must be increased to improve resolution. But increasing f_0 will increase attenuation as well, as discussed in Subsection 4.4. The consequence is that centre frequency and resolution size is always traded off.

This topic is treated again in the chapter on image quality in this web book.

8 Spatial compounding

The array technique described in Subsection 6 can be used to implement so-called spatial compounding. In this technique, several images are recorded from different angles and then combined, to yield an image with some desirable properties, relative to the conventional B-mode image. The technique is illustrated in Figure 8. Because a single compound image consists of N_θ single-angle images, the frame-rate will be reduced by a factor of N_θ compared to B-mode imaging.

An example of a conventional B-mode image and the corresponding compound image is shown in Figure 9. If compared to the B-mode image, a number of (desirable) features becomes apparent:

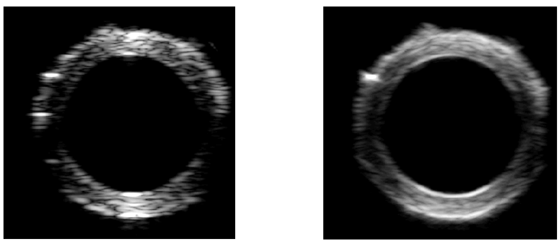

Figure 9 Left: Conventional image of a porcine artery. Right: Spatial compound image of the same porcine artery (average image of single-angle images from the angles: -21°, -14°, -7°, 0°, 7°, 14°, 21°).

The B-mode image has a quite "mottled" appearance, in the sense that the image consists of dots - roughly the size of the psf - on a black background. This is the result of the before mentioned constructive and destructive interference from closely spaced scatterers and reflectors, as illustrated in Figure 5. The phenomenon is commonly referred to as speckle noise. Speckle noise is a random phenomenon, and a given combination of constructive and destructive interference from a cloud of closely spaced scatterers is closely related to beam size, shape, orientation and direction. Thus, the interference pattern will change for the same tissue region when imaged from a different direction. If the change in view-angle is large enough, this interference patterns will be uncorrelated; so averaging of several uncorrelated single-angle images, will yield a reduction in speckle noise.

Because the ultrasonic echoes from interfaces vary in strength with the angle of incidence, the more scan angles used, the larger the probability that an ultrasound beam is perpendicular or nearly perpendicular to an interface, and the better the interface will be visualized.

The reduction in speckle noise and the improvement in visualization of interfaces gives an image with a more smooth appearance, better contrast and better delineation of boundaries. This can be seen in Figure 9 (right).

9 Case problems

9.1 Huygens' principle

How would you go about figuring out how many point sources are needed on the transducer surface, in Figure 3 in order to represent the pressure field in front of the transducer with a given accuracy?

9.2 Being critical about what you just read

The received signal in Figure 5 is not totally complete. Find the missing component(s).

In Figure 9 left, there is two bright dots at 9 o'clock and 10 o'clock, but only one at 10 o'clock in Figure 9 right. Why?

10 Nomenclature

a	Radius of transducer disk (m)
R	Radius of curvature of spherically focused transducer (m)
λ	Wavelength of ultrasound (m)
f_0	Centre frequency of emitted pulse (Hz)
c	Propagation speed of ultrasound (m/s)
N_θ	Number of single-angle images in spatial compound ultrasound
N_l	Number of scan lines in an ultrasound image
$f_r = T_r^{-1}$	Pulse repetition frequency (Hz)
D_{max}	Maximal depth (m)
$g_r(t)$	Received signal (V)
$\lvert\tilde{g}_r(t)\rvert$	Envelope of received signal (V)
α	Attenuation (m^{-1})

p Pressure (Pa)

z Specific acoustic impedance (rayl = kg/(m^{2}s))

ρ Physical density of medium (kg/m^3)

κ Compressibility of a medium (Pa^{-1})

11 Glossary

Refraction "The deviation of light in passing obliquely from one medium to another of different density. The deviation occurs at the surface of junction of the two media, which is known as the refracting surface. The ray before refraction is called incident ray; after refraction it is the refracted ray. The point of junction of the incident and the refracted ray is known as the point of incidence. [...]".[1]

Isotropic "Similar in all directions with respect to a property, as in a cubic crystal or a piece of glass."[1]

dB A magnitude variable, such as pressure, p, in Pa, can be written in as $20\log_{10}(p/p_{\mathrm{ref}})$ dB, where p_{ref} is some given reference pressure, needed to render the argument to the logarithm dimension less. Likewise intensities, I, can be written as: $10\log_{10}(I/I_{\mathrm{ref}})$ dB.

12 References

[1] Dorland's Illustrated Medical Dictionary. 27th edition. W. B. Saunders Co., Philadelphia, PA, USA. 1988.

[2] Kinsler LE, Frey AR, Coppens AB & Sanders JV: Fundamentals of acoustics. 3rd ed. John Wiley & sons, Inc. New York, NY, USA, 1982.

[3] Orofino, DP: Analysis of angle dependent spectral distortion in pulse-echo ultrasound. PhD dissertation, Department of Electrical Engineering, Worcester Polytechnic Institute, August 1992, USA.

[4] Kino, GS: Acoustic waves. Prentice-Hall, Inc. Englewood Cliffs, New Jersey, USA. 1987.