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3rd year Licence (LMD)

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Preface

This handout of the acoustics course was written for 3rd year students in the Physics of Materials License who are preparing, as part of the L.M.D. reform, a License in the field of "Material Sciences". It complies with the official program. It was written with the aim of providing a work and reference tool covering the knowledge required of them.

The manuscript is composed of five chapters, the first is mainly devoted to reminders on oscillations and resonance. The second chapter is reserved for understanding sound and sound sources. The properties of the acoustic wave are detailed in the third chapter. Medical diagnosis by ultrasound was studied in the fourth chapter. The last chapter is devoted to the presentation of the role of sound waves in prospecting and industry.

Although this manuscript has been prepared with the greatest care, it obviously remains improvable. I thank in advance anyone who sends me remarks or comments.

Zakaria Hadef

January 2025

Contents

Chapter-I:

Reminders on Oscillations & Resonance

I.1. Definition

I.1.1 Oscillation (Vibration)

Vibration is an oscillatory physical phenomenon of a body moving around its equilibrium position.

I.1.2 Periodic movement

A periodic movement is a movement which repeats itself and each cycle of which reproduces itself identically. The duration of a cycle is called period T which is expressed in seconds (s).

- The number of repetitions per second is called frequency (noted f, measured in (Hertz) or (s-1)). It is linked to the period by: f= 1/T.

- The number of revolutions per second is called pulsation (denoted ω, measured in rad/s), $\omega=2\pi f=2\pi/T$.

Mathematically, the periodic movement of period T is defined by: x (t +T) = x (t).

I.I.3 Vibratory movement

A vibrational movement is a periodic movement occurring on either side of an equilibrium position.

I.1.4. Sinusoidal movement

A vibratory movement is sinusoidal, if the elongation x (y or z) of a vibrating point is a simple sinusoidal function of time of type:

$x(t) = A\sin(\omega t + \varphi)$ Or $x(t) = A\cos(\omega t + \varphi)$

x(t) is called the elongation (or position) at time t.

A: the amplitude of a movement or maximum elongation.

ω: The pulsation of the movement and expressed in (rad/s).

φ: the initial phase corresponds to the phase at time t = 0

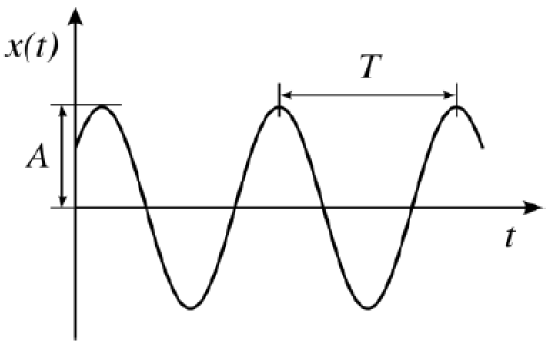


Figure I.1: Example of a sinusoidal periodic movement

I.1.5 Oscillatory Movement

An oscillating system is characterized by periodic movements near an equilibrium position under the effect of an external disturbance. When the movement is sinusoidal, the oscillator is said to be harmonic.

There are two types of vibration types (oscillations):

- Mechanical oscillations (simple pendulum, vibrating rope, etc.)
- Electromagnetic oscillations (Light, radio waves, etc.)

I.2. Choice of method

The choice of the calculation method to use to arrive at the equation of motion (EDM), we distinguish:

- Newton's equation.
- Lagrange equation.

In the rest of our reminder we choose Newton's equation to solve the equation of motion.

I.2.1 Newton's equation:

This formalism is based on the fundamental principle of dynamics and is applied according to the case of movement, namely: translation or rotation.

I.2.1.1 Translation movement

If a system of mass m is subjected to external forces, the fundamental law of dynamics gives us:

$$\sum_i \vec{F}_i = m\vec{\gamma}_i = m\vec{a}_i$$

I.2.1.2 Rotational movement

The fundamental law of the dynamics of a rotational movement is written:

$$\sum_i M_\Delta\left(\vec{F}_i\right) = J\,\ddot{\Theta} = J\frac{d^2\Theta}{dt^2}$$

I.2.2 Lagrange equation:

The Lagrange function (Lagrangian of the system) which is the difference of the kinetic energy E_c and the potential energy E_P as can be seen in the following formula:

$$L = E_c - E_P$$

$$\frac{d}{dt}\left(\frac{\partial L}{\partial \dot{q}}\right) - \frac{\partial L}{\partial q} = F_{ext,q}$$

Where q is the generalized coordinate which characterizes the vibratory movement and $F_{ext,q}$ *are* the generalized external forces.

I.3. Free undamped oscillations of systems with one degree of freedom

A system oscillating in the absence of any excitation force (external forces) is called free (undamped free oscillator. The number of independent variables describing the system is called the degree of freedom.

In mechanics, we call a harmonic oscillator which, as soon as it is separated from its equilibrium position by a distance x (or angle θ), is subject to a restoring force opposite and proportional to the separation x (or θ).

The differential equation of an undamped free motion is of the form:

$$\ddot{q} + \omega_0^2 q = 0$$

The Lagrange equation for the free oscillations of a conservative system is given by:

$$\frac{d}{dt}\left(\frac{\partial L}{\partial \dot{q}}\right) - \frac{\partial L}{\partial q} = 0$$

The solution to the previous equation is given by:

$$q(\mathrm{t}) = \mathrm{A}\cos(\omega_0 \mathrm{t} + \varphi)$$

Where A is the amplitude of oscillations, φ is the initial phase, and ω_0 is the pulsation of the system, it depends on the constituent elements (mass, spring, wires, etc.).

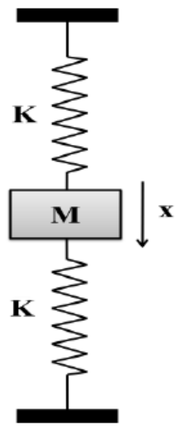
The constants A and φ are calculated from the initial conditions:

$$\begin{cases} q(t=0) = q_0 \\ \dot{q}(t=0) = \dot{q_0} \end{cases}$$

The amplitude of the oscillations of a free harmonic oscillator does not depend on time. Such oscillations are called undamped.

Example

Consider a system modelled by a mass **M** and two springs of stiffness *K*. (See the diagram opposite).



Numerical values: M = 0.5 kg, K = 100 N/m.

1- What is the type of system?
2- Establish the differential equation of motion as a function of *x*.
3- Give the final solution if: (0) = 1 *c* ;(0) = 0.

Solution

1- The type of system: Free system with one degree of freedom.

2- The movement equation is:

$$\sum \overrightarrow{F_{ext}} = \mathrm{M}\vec{a}$$

$$\mathrm{M}\ddot{\mathrm{x}} + 2\mathrm{kx} = 0 \Rightarrow \ddot{\mathrm{x}} + \frac{2\mathrm{k}}{\mathrm{M}}\mathrm{x} = 0$$

- The differential equation of motion is of the form:

$$\ddot{x} + \omega_0^2 x = 0$$

- The own pulsation:

$$\omega_0^2 = \frac{2K}{M} \Rightarrow \omega_0 = \sqrt{\frac{2K}{M}} = \sqrt{\frac{200}{0.5}} = 20 \text{ rad.s}^{-1}$$

- The proper period T_0:

$$T_0 = \frac{2\pi}{\omega_0} = 2\pi\sqrt{\frac{M}{2K}} = 2\pi\sqrt{\frac{0.5}{200}} = 0{,}314 \text{ s}$$

- The natural frequency f_0:

$$f_0 = \frac{\omega_0}{2\pi} = \frac{1}{2\pi}\sqrt{\frac{2K}{M}} = \frac{1}{2\pi}\sqrt{\frac{200}{0.5}} = 3{,}184 \text{ Hz}$$

3- The final solution is:

$$x(t) = A\cos(\omega_0 t + \varphi)$$

$$\dot{x}(t) = -A\omega_0 \sin(\omega_0 t + \varphi)$$

$$\begin{cases} \dot{x}(0) = -A\omega_0 \sin(\varphi) = 0 \Rightarrow \sin(\varphi) = 0 \Rightarrow \varphi = 0 \\ x(0) = A = 1 \text{ cm} = 0.01\text{m} \end{cases}$$

So, $x(t) = 0{,}01 \cos(\omega_0 t) \text{ m} \Rightarrow x(t) = 0{,}01 \cos(20t) \text{ m}$

I.4 Free damped oscillations of systems with one degree of freedom

I.4.1 Introduction

In damped oscillations, friction forces are taken into consideration. Friction is viscous and depends on the speed.

I.4.2 Damped oscillator

Harmonic oscillatory motion does not exist in reality: any real mechanical system exhibiting friction, part of the mechanical energy is dissipated as heat, and the oscillations of the system are damped.

I.4.3 Friction and damping coefficient

I.4.3.1 Viscous friction

The viscous friction force is proportional to the speed of movement and in the opposite direction.

$$\vec{F} = -\alpha . \vec{v}$$

Where α is the coefficient of viscous friction (kg/s).

I.4.3.2 Solid friction

The friction force is constant and opposite of the motion.

I.4.4 Lagrange equation

The Lagrange equation in the case of a damped free system is written:

$$\frac{d}{dt}\left(\frac{\partial L}{\partial \dot{q}}\right) - \frac{\partial L}{\partial q} + \frac{\partial D}{\partial \dot{q}} = 0$$

The dissipation function is defined by:

$$D = \frac{1}{2}\alpha.\dot{q}^2 \Rightarrow \frac{\partial D}{\partial \dot{q}} = \alpha.\dot{q}, \dot{q} = \dot{x}_\alpha$$

Wherex_α is the displacement of the shock absorber

The form of the differential equation is:

$$\ddot{q} + 2\delta\dot{q} + \omega_0^2 q = 0$$

Where δ is the damping coefficient and ω_0 is the natural pulsation.

I.4.5 Damped oscillator regimes

The differential equation of a damped oscillator is: $\ddot{q} + 2\delta\dot{q} + \omega02q = 0$. There are three possible regimes:

I.4.5.1 Aperiodic regime ($\delta > \omega_0$)

The friction is significant, the value of the damping coefficient is large, the system slowly returns to its equilibrium position without oscillating and the solution of the differential equation is of the form:

$$q(t) = A_1 e^{s_1 t} + A_2 e^{s_2 t}$$

$$q(t) = A_1 e^{\left[-\delta - \sqrt{\delta^2 - \omega_0{}^2}\right]t} + A_2 e^{\left[-\delta + \sqrt{\delta^2 - \omega_0{}^2}\right]t}$$

Where A_1 and A_2 are integration constants defined by the initial conditions.

I.4.5.2 Critical regime ($\delta = \omega_0$)

The return to the equilibrium position is done without oscillation and quickly. It allows to fix the limit between the pseudo-periodic and aperiodic regimes, the solution is of the form:

$$q(t) = (A_1 + A_2)e^{-\delta t}$$

Where A_1 and A_2 are integration constants defined by the initial conditions.

I.4.5.3 Pseudo-periodic regime ($\delta < \omega 0$)

Regime of low damping in which we observe that the amplitude of the oscillations is however decreasing, it is weighted over time by a decreasing exponential factor depending on the friction, and the solution of the differential equation is of the form:

$$q(t) = Ae^{-\delta t} cos(\omega t + \varphi)$$

Where A, φ are integration constants determined from the initial conditions, and ω is the pseudo pulsation defined by:

$$\omega = \sqrt{\omega_0^2 - \delta^2}$$

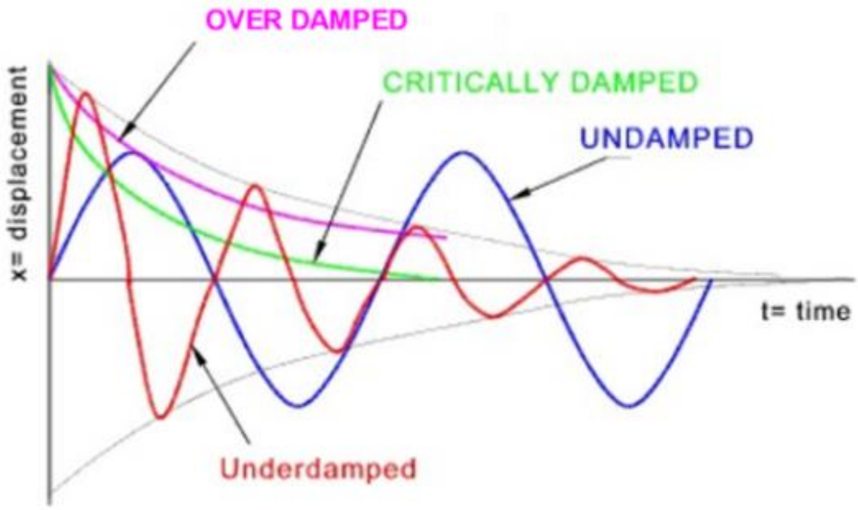


Figure I.2: The variation of q(t) as a function of time t for the three regimes (aperiodic regime, critical regime and pseudo-periodic regime).

I.4.6 Logarithmic decrement

The logarithmic decrement is the logarithmic measure of the periodic decrease of an oscillatory quantity (amplitude of the movement).

$$D = \frac{1}{n}\ln\frac{A(t)}{A(t+nT)}$$

Where n is the number of periods

By replacing the amplitude formulas, we obtain:

$$D = \delta\, T$$

Where δ is the damping coefficient and T is the pseudo period. It is given by:

$$T = 2\pi/\omega$$

ω is the pseudo-pulsation, it is equal to:

$$\omega = \sqrt{\omega_0^2 - \delta^2}$$

I.5. Forced oscillations of systems with one degree of freedom

I.5.1 Introduction

According to the previous part, it was found that damping reduces the amplitude of vibration, and the damping of oscillations was due to a decrease in mechanical energy, to overcome the friction responsible for energy losses and slowdowns of moving systems, it is necessary to apply an external force called excitation.

I.5.2 Differential equation of motion

The differential equation of forced oscillations of systems with one degree of freedom given by: is

$$\frac{d}{dt}\left(\frac{\partial L}{\partial \dot{q}}\right) - \frac{\partial L}{\partial q} + \frac{\partial D}{\partial \dot{q}} = F_{ext}$$

Where F_{ext} is the force generalized to an external force

The differential equation of the forced vibratory motion is:

$$\ddot{q}+2\delta\dot{q}+\omega_0^2 q=(t)$$

I.5.3 Solving the differential equation of motion

Figure I.3 presented the recording of the motion of a forced and amortized oscillator we notice two regimes: the first is short is called transient and the second is long is called permanent. So the solution of the equation of motion consists of two solutions.

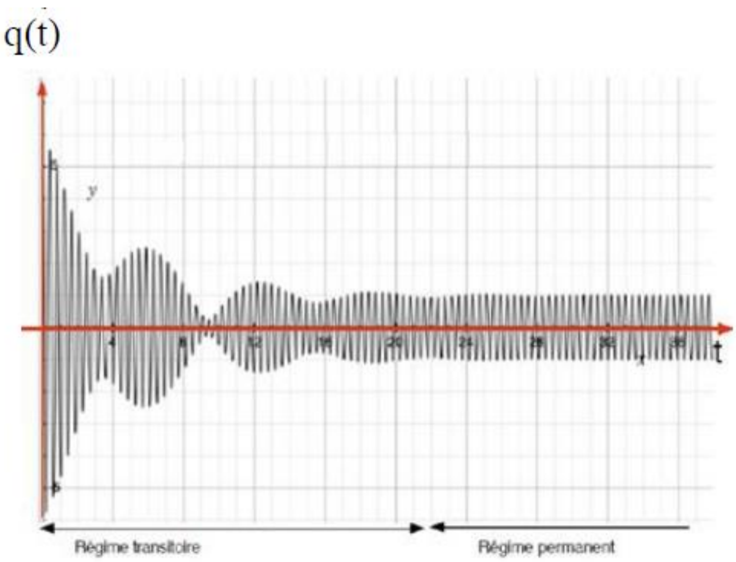


Figure I.3: Recording the forced and amorted motion oscillator.

The differential equation obtained is therefore a second order linear equation with constant coefficients with second member having the solution q(t). The solution q(t) of the differential equation which presents the response of the system to the external force is the sum of two terms:

$$q(t) = q_T(t) + q_P(t)$$

Where $q_T(t)$ and $q_P(t)$ represent respectively the general solution of the homogeneous equation and the particular solution.

- $q_T(t)$ is the (transient) solution of the homogeneous equation (without F). It is called transient because it goes out over time.
- $q_P(t)$ is the (permanent) solution of the non-homogeneous equation (with). It is called permanent because it lasts throughout the movement.

For sinusoidal excitation of type

$$F(t)=f_0 cos(\omega t) = f_0 e^{j\omega t}$$

The particular solution to the following form:

$$q_P(t) = Acos(\omega t+\varphi)$$

Where A is the amplitude, φ the phase shift of the total solution. or $q(t)$ is written in the following complex form:

$$q_P(t) = Ae^{j(\omega t+\varphi)}$$

We replace in the differential equation of motion.

$$\ddot{q} + 2\delta\dot{q} + \omega_0^2 q = (t) \ (4.6)$$

We have:

$$\dot{q} = Aj\omega e^{(\omega t+\varphi)}$$

And

$$\ddot{q} = -A\omega^2 e^{(\omega t+\varphi)}$$

So: $-A\omega^2 e^{(\omega t+\varphi)} + 2\delta\ Aj\omega e^{j(\omega t+\varphi)} + \omega_0^2\ Ae^{j(\omega t+\varphi)}$

$= f_0 e\ j\omega t\ [(\omega^2 - \omega_0^2) + 2\delta\ Aj\omega]^{\ j\varphi} = f_0$

By comparing the two sides of the equation we obtain

$$A = \frac{f_0}{\sqrt{(\omega^2-\omega_0^2)^2+4\delta^2\omega^2}}$$

*

And

$$\varphi = Arctg\frac{2\delta\omega}{\omega^2-\omega_0^2}$$

1.6. Resonance pulse:

The excitation pulse ω for which the amplitude A reaches its maximum is called the resonance pulse ωr. A is maximum when

$$\frac{\partial A}{\partial \omega}=0$$

$$\frac{\partial A}{\partial \omega} = \frac{\partial}{\partial \omega}\frac{f_0}{\sqrt{(\omega^2-\omega_0^2)^2+4\delta^2\omega^2}}$$

$$=\frac{f_0\left(4\omega\left(\omega^2-\omega_0^2\right)+8\delta^2\omega\right)}{2\sqrt{(\omega^2-\omega_0^2)^2+4\delta^2\omega^2}}$$

$$\frac{\partial A}{\partial \omega}=0 \Rightarrow f_0\left(4\omega\left(\omega^2-\omega_0^2\right)+8\delta^2\omega\right)=0$$

$$\Rightarrow \begin{cases} \omega \equiv \omega_r = 0 \\ ou \\ \omega \equiv \omega_r = \sqrt{\omega_0^2-2\delta^2} \end{cases}$$

So the maximal amplitude is :

$$A_{max} = \frac{f_0}{\sqrt{4\delta^2\omega^2-4\delta^4}}$$

For there to be resonance, it is necessary that:

$$\omega_0^2-2\delta^2>0 \Rightarrow 1-\frac{1}{2Q^2}>0$$

That means that the quality factor is $Q > 1/\sqrt{2}$ ⇒ Low damping.

The variation of the amplitude A and the phase shift φ are presented in the following figures

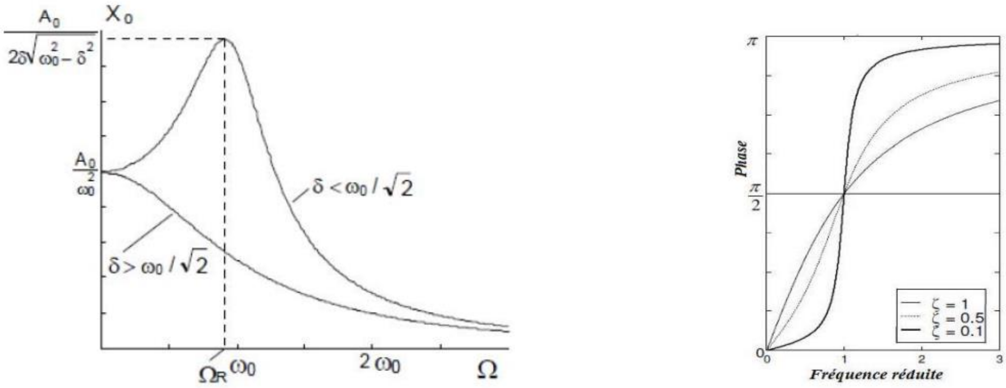


Figure I.4: Evolution of the (a) amplitude (b) phase as a function of pulsation

Chapter-II:

Sound & Sound Sources

II.1. Basic notions

II.1.1 Definition

II.1.1.1 Acoustic

Etymologically speaking, the word "Acoustics" comes from the Greek and means to hear. Acoustics is the science or let us say, a part of the physical sciences relating to the study of sounds and noises (everything related to sounds). It studies sound from the point of view of their emission, propagation and reception (Figure II.1).

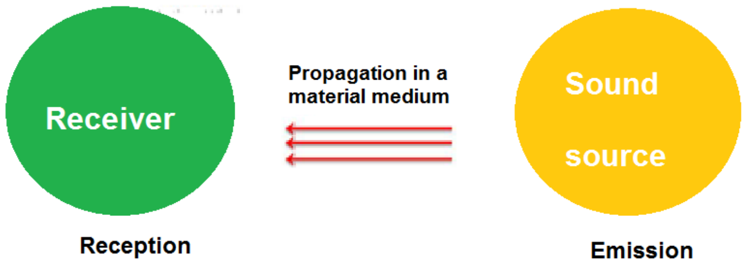


Figure II.1: Study areas of acoustics

II.1.1.2 Sound

Sound ''acoustic wave'' is a mechanical wave that requires an elastic and deformable material medium to propagate. It does not propagate in a vacuum. The sound wave consists of a propagation from near to near of a deformation, which locally causes pressure variations and oscillations of molecules around their equilibrium position, the sound wave is longitudinal. Sound waves are emitted by a source (human voice, musical instrument, tuning fork) and highlighted by a receiver such as the human or animal ear, a sound level meter. They are characterized by their frequency. They are divided into three bands (Figure II.2).

Figure II. 2: Sound Intervals.

Infrasound has a frequency lower than 20 Hz, and has the ability due to its low frequency to propagate over great distances in the air as well as in water or through the ground. They are audible to some animals such as elephants which allow them to communicate with each other up to 10 km. Sounds audible to the human ear are between 20 Hz and 20,000 Hz. Low frequency sounds are bass sounds, high-pitched sounds are high frequency sounds. Ultrasounds have frequencies higher than 20,000 Hz. They are audible to bats, cats, dolphins, etc.

II.1.1.3 Sound source

It can be said "sound source", any subject emitting sound waves: a speaking man, walking machine, speaker, foliage in front of the wind etc. There are two types of sound sources: Punctual sources and linear sources.

a. Punctual source

This is any source whose dimensions are negligiblely small compared to the propagation distances of waves such as: a musical instrument, a car in walking, a speaker etc. It can be fixed (musical instrument) or mobile (walking car). Punctual waves produce spherical waves in the atmosphere (Figure II.3.a).

b. Linear source

This is any source consisting of several punctual sources that are stored online (a running car line) (Figure II.3.b).

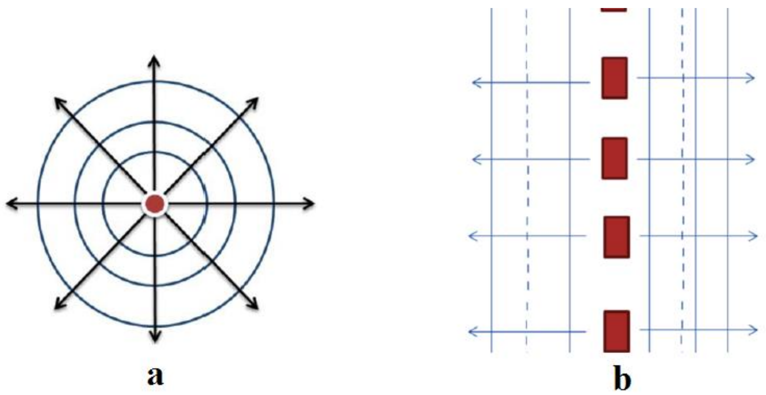


Figure II.3: Propagation of a sound wave: (a) punctual, (b) linear.

It is important to mention that linear waves produce cylindrical waves.

There are also two types of sound propagation in the air: omnidirectional propagation and directional propagation. The first occurs in the open air without the presence of any obstacle (Figure II.4.a). On the other hand, directional propagation occurs in an air limited by one or more obstacles (Figure II.4.b & II.4.c). The obstacles could be the floor, the walls, the objects etc.

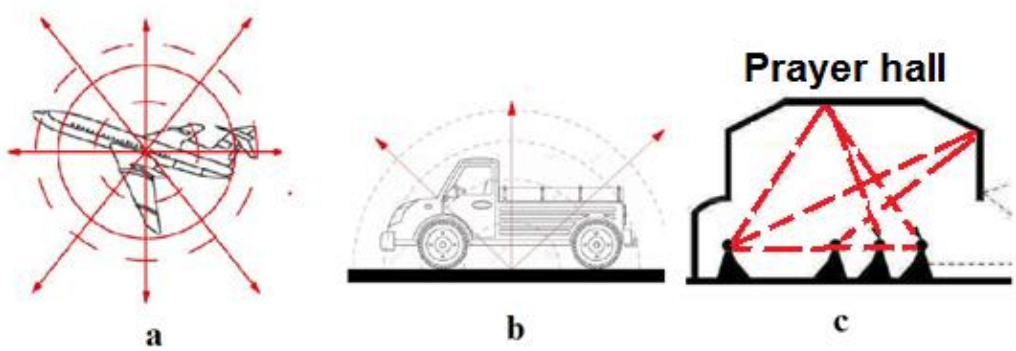


Figure II.4 : Omnidirectional and directional propagations: (a) sound wave generated by an plane which flies, (b) sound wave generated by a car in which the obstacle is the soil, (c) sound wave generated by an Imam who gives a lesson to his worshipers in a Prayer room whose obstacles are the walls.

II.1.2 Sound Types

II.1.2.1 Pure Sound

Pure sound is an auditory sensation caused by a purely sinusoidal periodic pressure wave. It is only composed of a single frequency (Figure I.5).

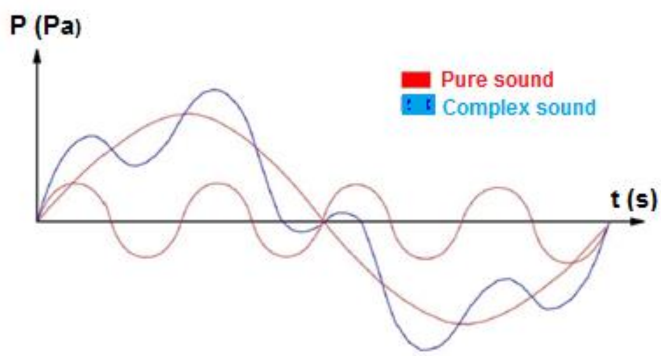


Figure II.5 : Pure & complex sounds

II.1.2.2 Complex Sound

Complex sound is a mixture of pure sounds. In other words, it is the superposition of a certain number of pure sounds (Figure II.5). Complex sounds are natural sounds; they have several pure sounds, which can be separated during a spectral analysis.

II.1.2.3 Confused sound

It is a mixture of sounds without precise periodicity like the rustles of the leaves in the trees (Figure II.6).

Figure II. 6: Confused sound.

II.1.2.4 Musical Sound

Music is the art of writing a piece using the elements of melody, harmony, rhythm, and timbre to arrange sounds in time. It is one of the universal cultural components of all human cultures.

Pitch (which governs melody and harmony), rhythm (and its related concepts of tempo, meter, and articulation), dynamics (loudness and softness), and timbre and texture are all common elements in general descriptions of music, Any of these elements can be highlighted, de-emphasized, or omitted depending on the style or genre of music.

A good, continuous, and uniform sound created by normal and periodic vibrations, such as that produced by a violin, flute, tuning fork, and so on, is known as musical sound. Noise is a loud, unpleasant sound that is caused by an irregular series of disturbances and is discontinuous, such as the sound of a gunshot, cracker bomb, or another explosive device.

II.1.3 Noise

II.1.3.1 Definition

Noise is a sound or a set of annoying sounds. This is a harmful signal for humans, annoying their normal activity. Noise is a mixture of sounds, several frequencies, of several levels.

Noise could be the origin of some problems that are:

1. Deafness: if its level exceeds critical value.
2. Reduction in work yield
3. The noise can be simply boring and thus provoking the loss of sleep, nervous crises and stress.

II.1.3.2 Tinnitus and hyperacusis

Tinnitus is an abnormal auditory sensation that only the affected person perceives. In other words, the person with the symptom of tinnitus occasionally or continuously hears whistling without external sound stimulation.

Hypoacusis is a disorder that affects hearing. In this case, the affected person becomes hypersensitive to sounds that are usually tolerated by the unaffected person. In other words, the noise tolerance threshold of this affected person is lower.

II.1.3.3 Background noise

Background noise refers to the parasitic signals existing during a period. It is a multitude of frequencies that will be added to a sound signal. It can be generated by wind, conversations, machines or equipment, traffic, sounds coming from corridors, other rooms, etc.

II.1.3.4 Ambient noise

Ambient noise is defined as all surrounding noise. At the moment when there is a particular sound source, ambient noise includes the noise emitted by this source and other residual noises. In short, it refers to all noises emitted by all nearby or distant sources.

II.1.3.5 Residual noise

Residual noise is the sound level in the absence of any particular noise generated, for example by a machine.

II.1.3.6 Emergence noise

Emergence is the difference between ambient noise (containing the particular noise) and residual noise composed of usual exterior or interior noises.

II.1.3.7 Noise measurement

The sound level meter is a device for measuring sound pressure (Figure II.7). It consists of a microphone that responds to sound pressure, a preamplifier of filters, an amplifier and a galvanometer. Today, sound level meters are more developed. This device allows measuring the overall linear pressure level, the unweighted pressure level per octave, the weighted overall pressure level, the weighted pressure level per octave and the reverberation time.

Figure II.7 Sound level meter

II.2. Waves generation & waves source<

II.2.1 Definition

Wave is defined as a disturbance that propagates. There are two types: mechanical waves and electromagnetic waves.

II.2.1.1 Mechanical waves

Mechanical wave is a disturbance or oscillation that propagates through a medium (air, a rope, water, earth). It is a displacement of energy that occurs from near to far without transport of matter.

II.2.1.2 Electromagnetic waves

Electromagnetic waves (light) do not need a material medium to propagate. They can propagate in a vacuum (radio communications in space, light from stars, etc.).

II.2.2 Sound wave

The sound wave is produced by the vibration of an object (the vocal cords of a Quran Reader, the membrane of a loudspeaker, the skin of a drum, the rustling of the leaves of a tree, etc.) (Figure II.8). As it is a mechanical interaction between two structures: hand and skin of a drum, finger and string of a guitar, etc. This vibratory movement propagates in a material medium, between the source and the ear of the listener. The molecules of the material medium are alternately compressed and dilated, giving rise to a sound wave. The propagation of sound occurs without transport of matter: the sound wave coming from a loudspeaker does not create an air current.

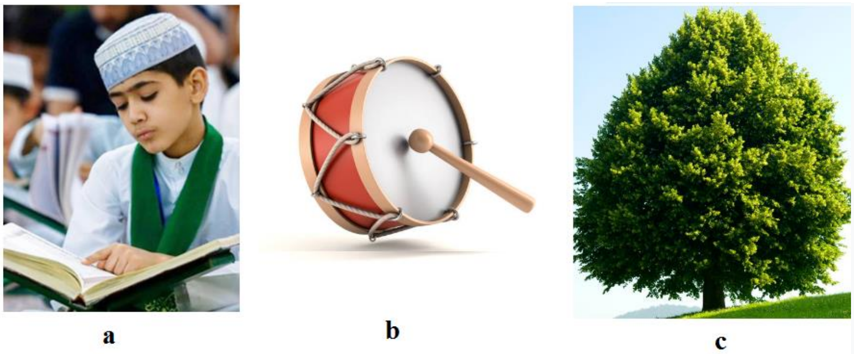


Figure II.8: Some sound sources: (a) the vocal cords of a reader from Coron, (b) the skin of a drum, (c) the rustling of the leaves of a tree.

The propagation of the vibratory movement corresponds to a transport of energy and not of matter (molecules). Each molecule composing the air in contact with a sound source vibrates in turn. This vibration is then transmitted to the neighboring molecule, then to the neighbor of this neighbor, etc. (propagation from near to near).

The sound wave is a longitudinal mechanical wave because the molecules move parallel to the direction of propagation of t wave (Figure II. 9). Consider a tuning fork that is producing sound waves. Just before the prongs of the tuning fork set into vibration, the medium is undisturbed.

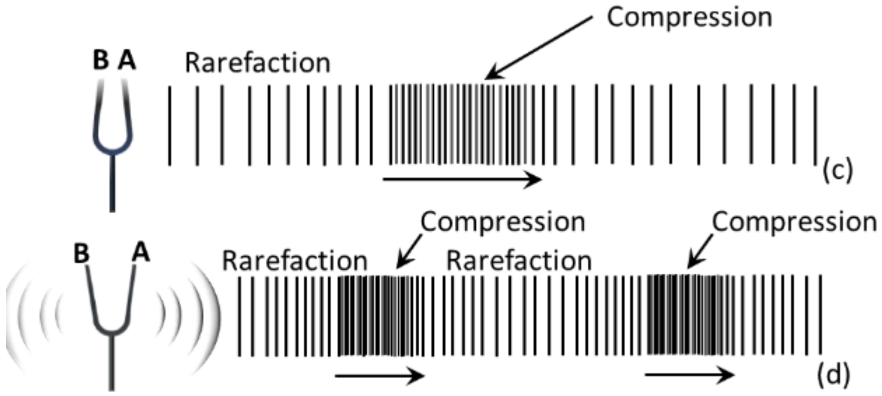


Figure II.10: The atoms move parallel to the direction of propagation of the sound wave generated by a tuning fork.

When the prong moves outward towards the right, it causes disturbance in the medium that compresses the air in front of it, causing the pressure to rise slightly. The region of increased pressure is known as a compression pulse, and it travels away from the prong with the speed of sound. After producing the compression pulse, the prong reverses its motion and moves inward. This drags away some air from the region in front of it, causing the pressure to dip slightly below the average pressure. This region of decreased pressure is known as a rarefaction pulse. Immediately the compression pulse and the rarefaction pulse travels away from the prong with the speed of sound. This keeps on happening because of which sound moves ahead.

II.3. Ultrasonic waves

Ultrasound is a mechanical wave traveling at a frequency above the human hearing range (f > 20 kHz). Ultrasonic waves can be transverse, longitudinal, Lamb or surface waves.

II.3.1 Transverse wave

In transverse waves, the displacement of the particle is perpendicular to the direction of propagation of the wave (Figure II.10).

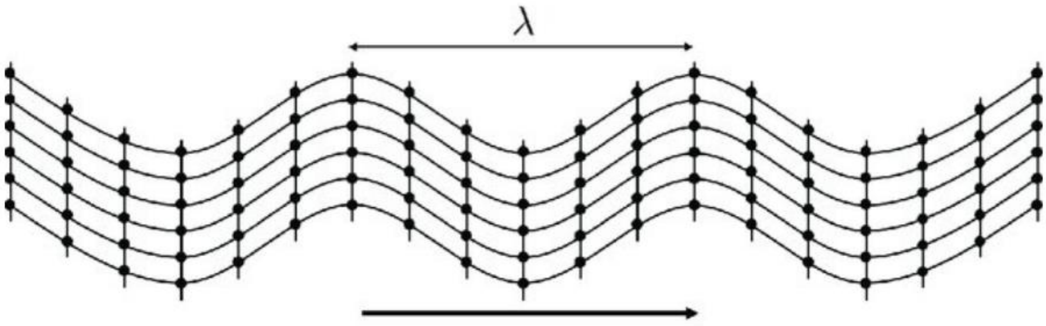


Figure II.10 : Transverse wave.

For a transverse wave to propagate, the medium must be sufficiently rigid. If we take the example of the rope (Figure II.11), the particles will oscillate vertically. In other words, they move alternately up and down.

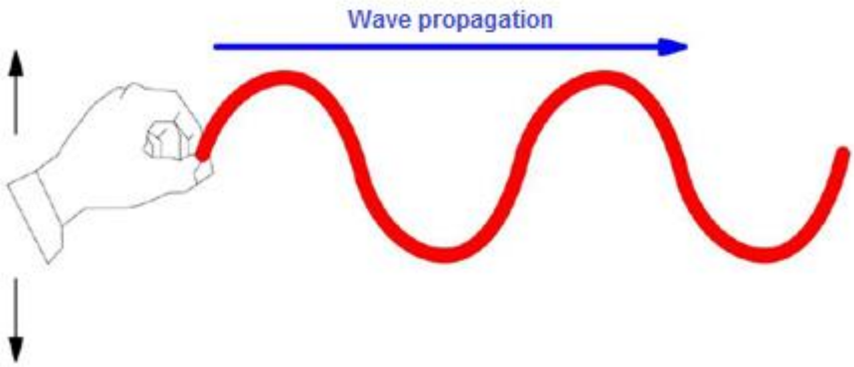


Figure II.11: Transverse wave generated by a string.

These waves only propagate in solid media or viscous liquid media; they will not propagate in gaseous or liquid media. Some examples of transverse waves are:

- The ripples on the surface of the water
- The secondary waves of an earthquake
- Electromagnetic waves
- The waves on a string
- Stadium or human wave
- The ocean waves

II.3.2 Longitudinal wave

In a longitudinal wave, the displacement of the particle is parallel to the direction of the wave propagation. (Figure II.12).

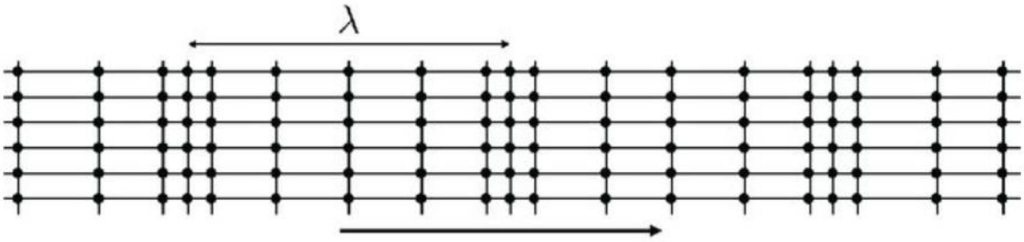


Figure II.12 : Longitudinal wave.

The example of a spring stretched horizontally then released can be cited. Each coil will move relative to its equilibrium position (Figure II.13).

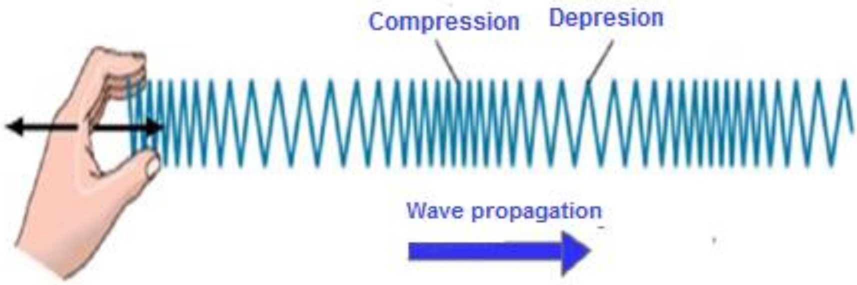


Figure II.13: Longitudinal wave generated by a spring.

Longitudinal waves propagate normally in gases, liquids and solids, however contained high frequencies in non-destructive testing "NDT". Some examples of longitudinal waves are:

- Sound waves in air
- The primary waves of an earthquake
- Ultrasound
- The vibration of a spring
- The fluctuations in a gas
- The tsunami wave

II.3.3 Surface waves

Surface waves travel more slowly through Earth material at the planet’s surface and are predominantly lower frequency than body waves. They are easily distinguished on a seismogram. Shallow earthquakes produce stronger surface waves; the strength of the surface waves are reduced in deeper earthquakes.

There are several types of these waves for example: Stannly, Love, Rayleigh…etc.

II.3.3.1 Love Waves

One kind of surface wave is called a Love wave (Figure II.14), named after British mathematician A. E. H. Love, who worked out the mathematical model for this wave type in 1911. Love waves produce entirely horizontal motion. The amplitude is largest at the surface and diminishes with greater depth.

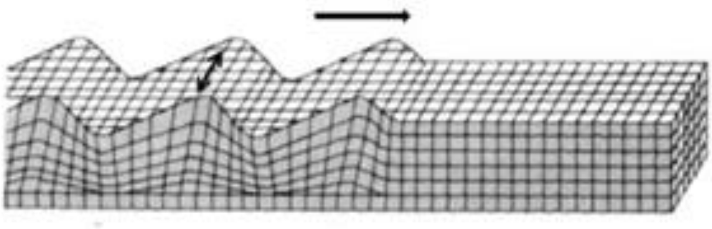


Figure II.14: Love Wave.

II.3.3.2 Rayleigh Waves

The other kind of surface wave is the Rayleigh wave (Figure II.15), named for John William Strutt, known as Lord Rayleigh, who mathematically predicted the existence of this kind of wave in 1885. A Rayleigh wave rolls along the ground with a more complex motion than Love waves. Although Rayleigh waves appear to roll like waves on an ocean, the particle motion is opposite of ocean waves. Because it rolls, it moves the ground up and down, and forward and backward in the direction that the wave is moving. Most of the shaking felt from an earthquake is due to the Rayleigh wave, which can be much larger than the other waves. Like Love waves, the amplitude of the wave decreases dramatically with depth.

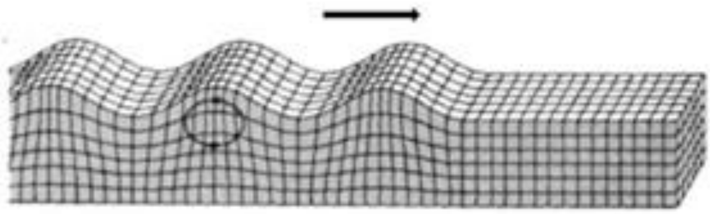


Figure II.15: Rayleigh Wave.

Chapter-III:

Acoustic Wave Properties

III.1. Celerity, Sound speed

III.1.1 Sound speed and its propagation medium

Sound speed depends on the medium in which the sound wave is propagated (air, water, earth, concrete, etc.) (Table III.1).

Table III.1: Sound speed in different propagation media.

Propagation medium	Sound speed (m/s)
Air	340
Water	1425
Steel	5000

Depending on the medium in which it is propagated, sound waves will propagate more or less quickly. In water, molecules are closer to each other than in air. Therefore, it is easier for a sound wave to propagate in water than in air (4 times faster in water than in air). In other words, the closer the molecules are, the faster the sound travels (Figure III.1).

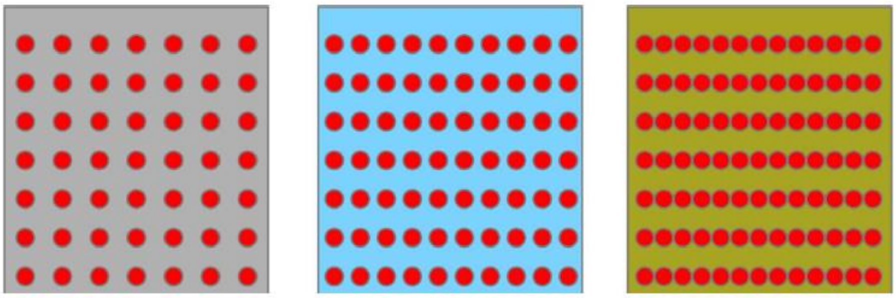


Figure III.1: Schematic representation of the molecules of the three propagation media (from left to right); air, water and steel

III.1.2 Relationship between sound speed and temperature

In air, the speed of sound (c) also depends on its temperature (t). The sound speed increases with increasing of temperature. It increases by 0.6 m/s for 1°C. It must not confuse the sound speed, which is the speed of propagation of waves in the air according to concentric spheres, and the particles speed that move very slightly around their equilibrium position. To calculate the speed of sound, we use the following formula

$$c = 20\sqrt{t}$$

Where (c) is the sound speed (m/s) and (t) is the air temperature (°C)

III.2. Acoustic pressure

Atmospheric pressure is the pressure exerted by air on all bodies. In the absence of a sound wave, airborne particles are distributed regularly, the pressure is equal to atmospheric pressure at all points. On the other hand, in the presence of a sound wave, there will be zones of pressure and depression (Figure III.2). Acoustic pressure is expressed in Pascals ($P_a = 1N/m^2$). Instantaneous pressure P is the sum of atmospheric or static pressure of the order of 105 (a variable value depending on climatic conditions) and acoustic pressure P_a.

$$\boldsymbol{P = P_0 + P_a}$$

Where P is the instantaneous pressure, P_0 is the acoustic pressure and P_a is the atmospheric pressure.

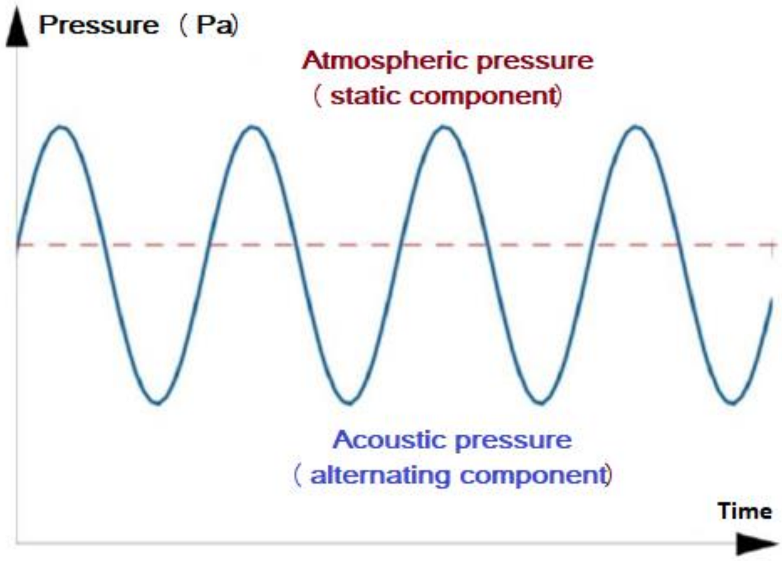


Figure III.2: Variation of pressure at a point in space subjected to an acoustic wave

III.3. Acoustic cavitation

The action of ultrasound in liquid media is based on the phenomenon of cavitation: creation, growth and implosion of bubbles formed when a liquid is subjected to a pressure wave (Figure III.3). This results in a pressure shock in the vicinity of the bubble and the formation of a thermochemical micro-reactor inside the bubble.

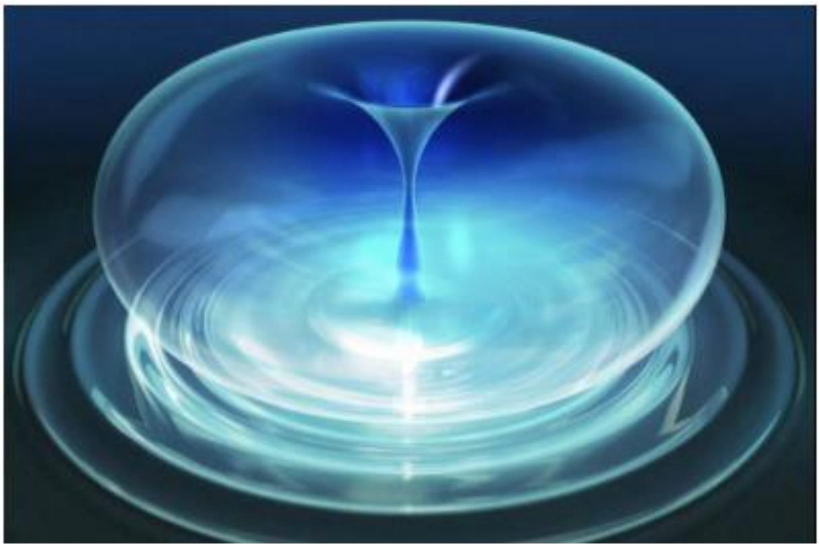


Figure III.3: Acoustic cavitation phenomenon.

For acoustic cavitation to occur, a power threshold must be reached. This threshold is of the order of 0.5W/cm² at 20 kHz for water at atmospheric pressure and of the order of a few W/cm² for organic solvents. The amplitude of the vacuum to be provided to reach the cavitation threshold depends on several parameters: the higher the viscosity of the medium (and therefore the internal cohesion of the liquid), the more difficult cavitation is to obtain because the particles are more difficult to separate. In liquid media, acoustic cavitation is the origin of the effects of power ultrasound.

III.4. Sound pressure level

The human ear is sensitive to pressure variations ranging from 2.10^{-5} Pascal to 20 Pascal. (20 Pa is the pain threshold). To calculate the sound pressure level (L_p), the following formula is used:

$$L_p = 20 \log \left(\frac{P}{P_0}\right)$$

P: measured sound pressure.

P0: reference sound pressure.

The pressure level of the audibility threshold is 0 dB and the pain threshold is 120 dB. It should be noted that the intensity or volume of a sound depends on the sound pressure created by the sound source (Figure III.4). The greater the pressure, the higher the volume.

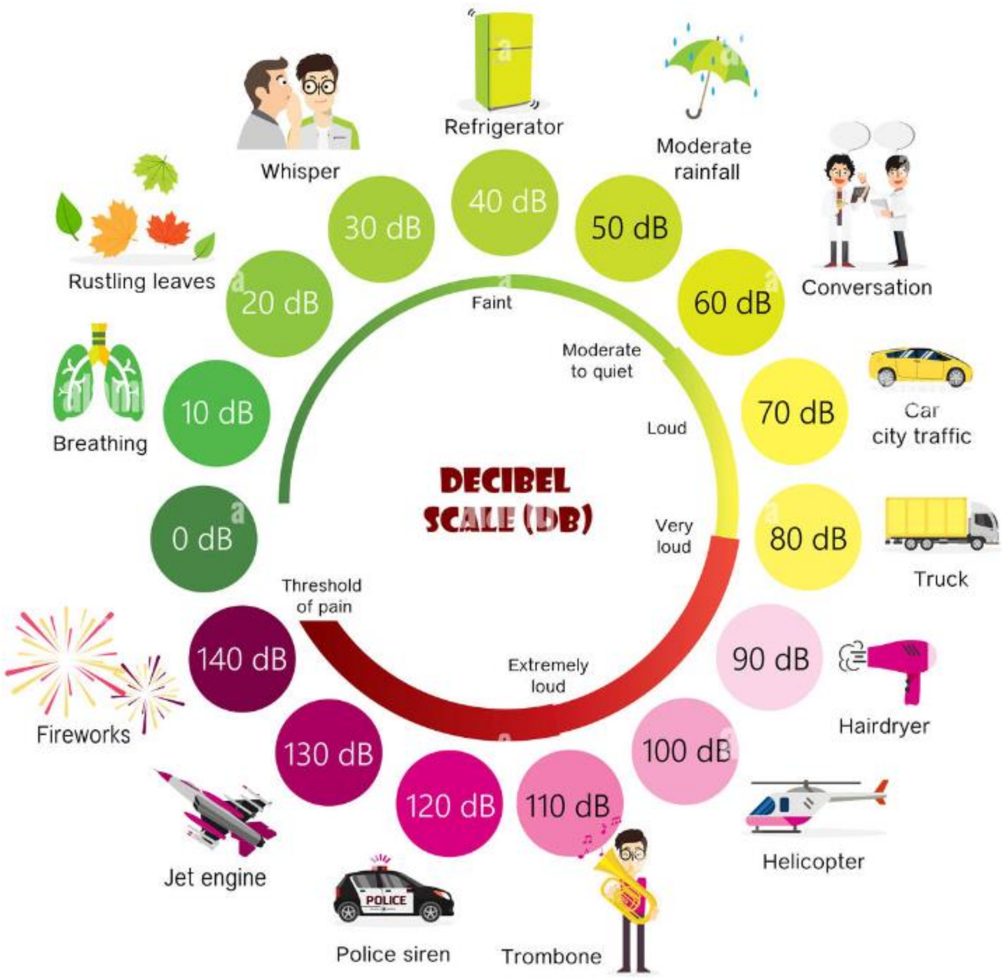


Figure III.4: Sound pressure level of some sound sources

III.5. Acoustic power and Sound intensity

III.5.1 Acoustic power

Acoustic power (*W*) is the energy released by a sound source per unit of time that is distributed in space to reach the ear. It is expressed in watts (*W*). The audibility threshold of a sound is 10^{-12} W at a frequency of 10,000 Hz.

The formula for calculating the acoustic power level is as follows:

$$L_W = 20 \log (W/W_0)$$

W: Measured acoustic power.

W_0: Reference acoustic power.

III.5.2 Sound intensity

Sound intensity is the quantity that informs us about the strength of a sound. Thus, the higher the sound intensity, the louder the sound perceived by the human ear. Sound intensity is expressed in watts per square meter ($W.m^{-2}$). There is a minimum sound intensity below which the sound is not heard; this is the threshold of audibility. It is worth $I_0 = 10^{-12}$ W.m-2.

$$I = P/S$$

I: Sound intensity

P: Acoustic power

S: Wave surface

The sound wave propagates in a spherical manner (in all directions) (Figure III.5). The surface area of a sphere is $4\,\pi\,r^2$, so the formula for calculating sound intensity is as follows

$$I = P/4\,\pi\,r^2$$

r : the distance between the sound source and where we are located

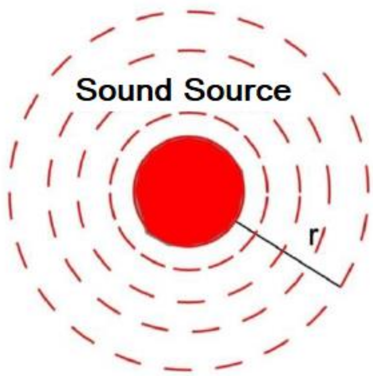


Figure III.5: Representative diagram of the surface of a sound wave

To calculate the sound intensity level (L_i), the following equation is used:

$$L_i = 10 \log (I/I_0)$$

L_i: the sound intensity level

I: the sound intensity

I_0: the reference sound intensity (audibility threshold ($I_0 = 10^{-12}$ $W.m^{-2}$)

III.6. Frequency and amplitude

III.6.1 Frequency

Frequency is the number of periods per second. In other words, it is the number of vibrations per second (Figure III.6). It is expressed in Hertz (Hz). For example, 200 Hz corresponds to 200 vibrations per second.

The period is the time taken by the air molecule to travel a single round trip (vibration for a round trip). It is the number of seconds per oscillation

$$f = 1/T$$

f: the number of vibrations per second

T: the number of seconds per vibration

III.6.2 Amplitude

The amplitude of a sound wave is the maximum displacement of an air molecule from its resting position (Figure III.6). The sound intensity that humans perceive depends on the amplitude. The greater the amplitude of a sound wave, the louder the sound. When you turn up the volume on your television, for example, you increase the amplitude of the sound wave coming from it.

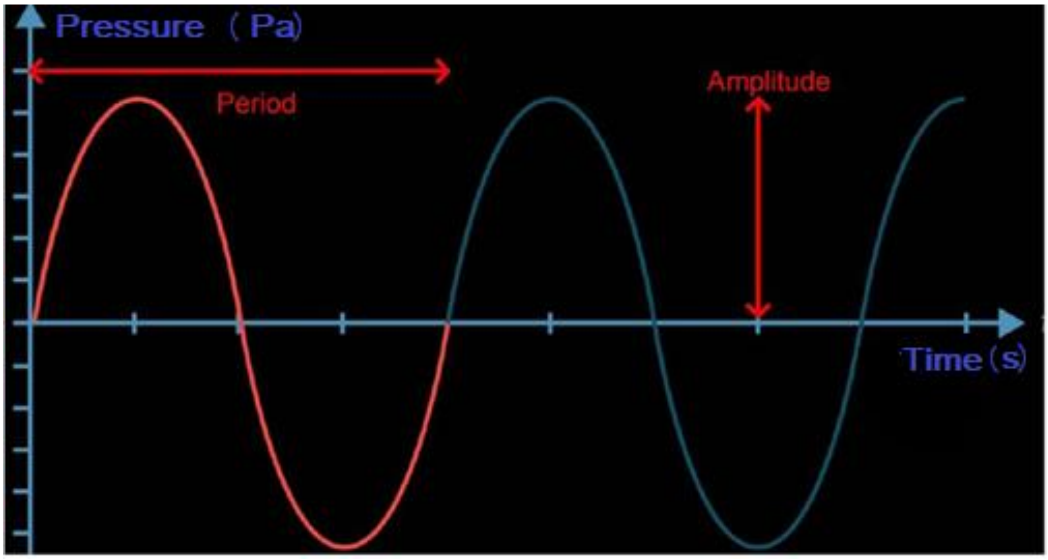


Figure III. 7: Figure III. 7: Oscillogram present the frequency a the nd amplitude of a sound wave

III.7 Acoustic impedance

Acoustic impedance (Z) expressed by $Kg/m^2.s$ or RAYL (Rayleigh)), it is a particularly important quantity to characterize a medium. It presents the resistance to wave propagation and is given by:

$$Z = \rho\, c$$

Table III.2: Some values of acoustic impedance.

Medium	Acoustic impedance Z ($10^6 Kg/m^2.s$)
Air	0.0004
Water	1,48
Liver	1,65
Bone	4,11-8
Blood	1,66
Fat	1,38
Brain	1,55
Muscle	1,65-1,74

III.8 Bel and decibel

The decibel is the dimensionless unit of measurement for sounds. It is used to measure sound intensity and other physical quantities. In fact, it expresses the ratio between the measured sound intensity and the reference sound intensity (the smallest intensity audible to humans). The decibel is composed of deci and bel, the bel is worth ten times the decibel. The Bel is named in honour of the sound inventor Graham Bell. In acoustics, the bel is the logarithm base 10 of the quotient of a measured physical quantity to a reference physical quantity.

The ear is a unique detector of sounds in that it captures sounds on a variety of extremely large quantities (Figure III.8). The logarithm allows us to translate our way of perceiving sounds. Consequently, we use the bel.

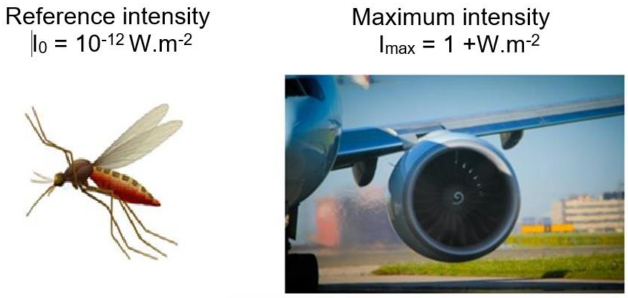


Figure III.8: Reference sound intensity and maximum sound intensity in humans.

To understand the principle, we can imagine a device that measures the size of an atom and the size of an Atomium. The first measures 1 Å, while the second measures 10^2 m. A device that measures such a variety of lengths does not exist (Figure III.9).

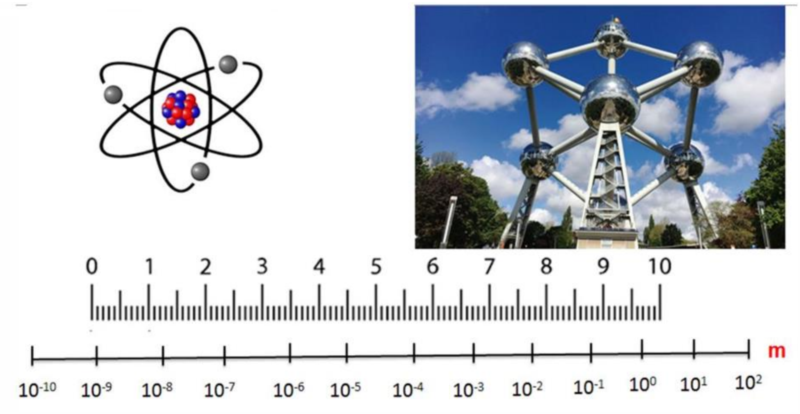


Figure III.9: The atom and the Atomium: two different sizes, two different scales.

For example, the slat does not adapt to the measurement of an atom, nor to the measurement of the Atomium. On the other hand, the ear can capture a great variety of sounds. To represent the measurement of the atom and that of the Atomium on the same scale, we use a logarithmic scale. The same principle is applied to sounds (Figure III.10).

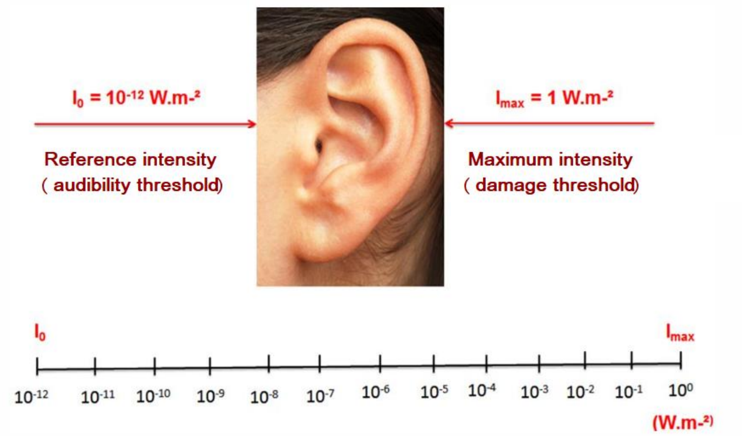


Figure III.10: The logarithmic scale applied to sounds

From the intensity to the decibel (Figure III.11), we have:

$I_0 = 10^{-12}$ W.m^{-2}

$I_{max} = 1$ W.m^{-2}

If we want to write I_{max} relative to I0, we will have:

$I_{max} = X\ I0$

$I_{max} = 10^{12} \times I_0 = 10^{12} \times 10^{-12} = 1.$

$I_0 = 10^{-12} = 10^0 \times 10^{-12} = 10^{-12}$

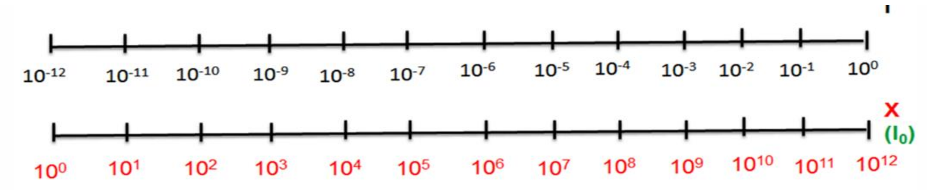


Figure III.11: I relative to I_0.

The relationship between physical intensity and the intensity (Figure III.10) perceived by the ear is:

$$y = Log\ (X)$$

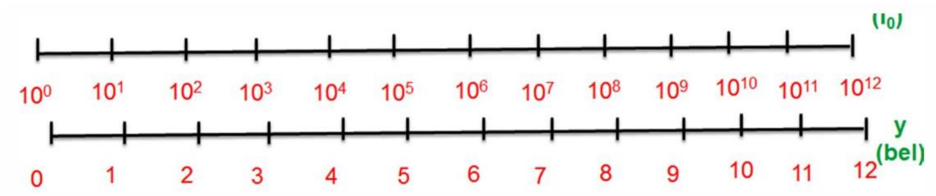


Figure III.12: From I^0 to bel.

1 bel = 10 decibel

y= 10 log(x) dB (Figure III.13)

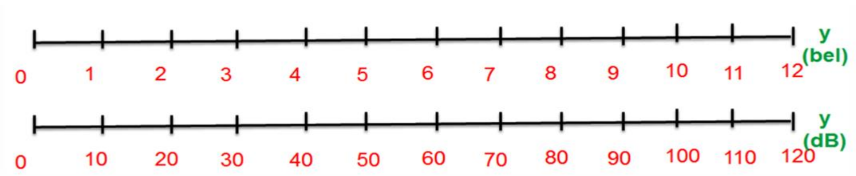


Figure III.13: From bel to decibel (dB).

III.9 Acoustic waves Interferences

A low intensity sound added to a high intensity sound does not change the latter. On the other hand, two sounds of the same intensity added to each other increase the level of one of them by 3dB. The important thing to remember is that we add the intensities and the sound powers but we never add the decibels

$$I_{Total} = \sum I$$

If we have *n* sources S_1, S_2… and S_n the superposition of *n* sound levels L_1, L_2… and L_n (the *n* sources operate simultaneously), emitted by the *n* sounds is carried out by the following formula:

$$L_r = 10 \log (10^{0,1L_1} + 10^{0,1\,L_2} + 10^{0,1\,L_n})$$

L_1 (dB): le niveau sonore en provenance de la première source S_1 fonctionnant seule

L_2 (dB): le niveau sonore en provenance de la première source S_2 fonctionnant seule

L_n (dB): le niveau sonore en provenance de la première source S_n fonctionnant seule

L_r (dB): le niveau sonore résultant au point de réception en provenance des deux sources sonores.

To subtract two sound levels, we use the following formula:

$$L_r = 10 \log (10^{0,1L_1} - 10^{0,1\,L_2})$$

Example :

Let's say there are four sound sources operating simultaneously with the following sound levels: L_1 = 40 dB, L_2 = 40dB, L_3 = 35 dB and L_4 = 47 dB.

The resulting sound level is equal to:

$L_r = 10 \log (10^4 + 10^4 + 10^{3,5} + 10^{4,7})$ dB

L_r = 47,38 dB

III.10 Ultrasound-Matter Interaction (Attenuation)

III.10.1 Definition

Attenuation is defined by the reduction in the intensity of an ultrasound beam during its propagation in a medium that depends on the nature and depth of the medium crossed and the beam frequency.

Ultrasound attenuation is due to the multiple interactions seen previously (absorption, reflection-refraction and diffusion) which reduce the intensity of the beam when it penetrates the tissues. This decrease obeys a decreasing exponential law:

$$I(x) = I_0 \exp (\mu x)$$

I: ultrasound intensity at distance x.

I_0: initial intensity.

μ: linear attenuation coefficient

x: the distance travelled by the wave from the source.

The attenuation coefficient expressed in decibels per unit of length crossed is defined by:

$$\boldsymbol{\alpha = 10\ \frac{\mu}{2.3}}$$

The wave frequency is not modified by the change of medium, but the speed changes and therefore the wavelength changes.

III.10.2 Attenuation by absorption

During wave propagation, part of the ultrasonic energy is absorbed by the medium

- The distance travelled by the wave (decreasing exponential law), the intensity decreases more quickly in the first cm.
- Frequency: the transfer of energy to the tissues increases with the frequency, the absorption coefficient α is given by the relation

$$\boldsymbol{\alpha = k\, f\, b}$$

Where k is a constant, f is the frequency of the acoustic wave and b is between 1 and 2 depending on the absorbing tissues. For water, $b = 2$. In general, for soft tissues, $b = 1$

It therefore appears that the higher the ultrasound frequency, the greater the absorption phenomenon (and therefore attenuation) and consequently the lower the maximum depth that can be visualized. In practice, the choice of frequency used is largely guided by the depth of the area that one wishes to explore.

- Low frequencies for exploring deep organs
- High frequencies for exploring superficial organs

III.10.3 Attenuation by Reflection and Transmission

When an acoustic wave encounters the interface separating two media of different acoustic impedances, part of the wave is transmitted into the other medium (it undergoes refraction) while another part is reflected on the interface (it undergoes reflection) (Figure III.14).

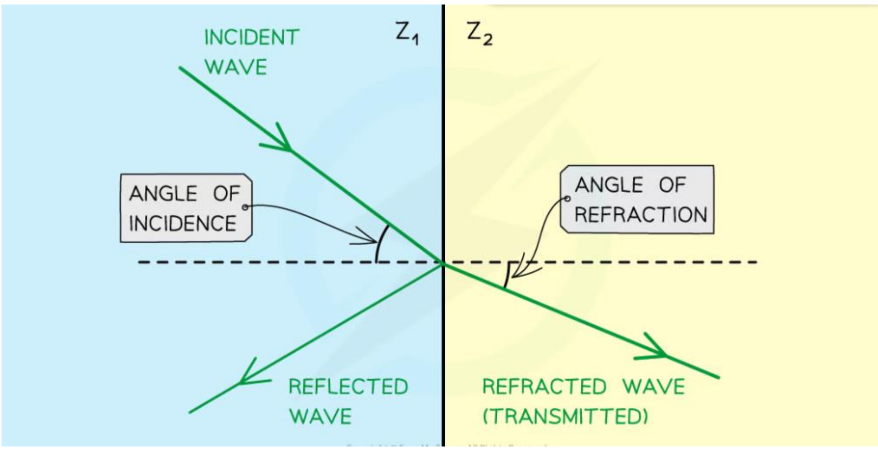


Figure III.14: Acoustic reflection and transmission.

The concept of acoustic impedance makes it possible to study this phenomenon completely and quantitatively, to estimate the quantities of acoustic energy transmitted and reflected

Like light rays, these two phenomena (reflection-refraction) obey Descartes laws, the different angles being measured relative to the normal of the interface: the amplitude of the signal received by the receiver is maximum when the angle of incidence equals the angle of reflection ($i_0 = i_r$), and the transmitted beam is deflected according to an angle it which depends on the celerities c_1 and c_2 of the wave before and after its passage through the interface (c_1 and c_2 being a function of the respective acoustic impedances Z_1 and Z_2 of media 1 and 2).

$$i_0 = i_r \quad ; \quad \frac{\sin i_0}{c_1} = \frac{\sin i_r}{c_2}$$

To do this, we study the ratio of sound intensities and then define two coefficients

- **Reflection coefficient** (Proportion of reflected ultrasounds):

$$R = \frac{I_{reflective}}{I_{incidental}} = \frac{Z_2 - Z_1}{Z_2 + Z_1}$$

- **Transmission coefficient** (Proportion of transmitted ultrasound):

$$T = \frac{I_{transmited}}{I_{incidental}} = \frac{4Z_2 Z_1}{Z_2 + Z_1}$$

We can see that R and T varied between 0 and 1, and R + T = 1.

Generally speaking, the impedances of the two media must be very different.

- If ***R = 0***, the acoustic adaptation appears: the wave is fully transmitted to medium 2. In this case $Z_1 = Z_2$.
- If ***R = 1***, the wave is fully reflected and does not pass into medium 2. In this case, one of the two impedances is negligible compared to the other.

The greater the difference in impedance between media 1 and 2, the greater the reflected energy and the greater the amplitude of the echoes, the more visible the defect will be.

III.10.4 Attenuation by diffusion

The expressions diffusion and scattering are often used interchangeably, but within acoustic terminology, they have different meanings.

Diffusion is an equal distribution of the different frequencies. But, Scattering disperses the sound in a non-specular direction, thus the distribution in unequal

Diffusion is definitely better in a listening room because an equal distribution colours the sound far less. Almost anything can scatter the sound, but the distribution will vary at different frequencies and at different angles. Thus the sound becomes collared.

The attenuation by diffusion occurs during the interaction of an acoustic wave of wavelength λ and:

- A small obstacle of dimension smaller than the wavelength ($d < \lambda$), this small object vibrates and re-emits in all directions of space a fraction of the energy contained in the incident wave.
- A rough and very irregular interface

Chapter-IV:

Ultrasound & Medical Diagnosis

IV.1. Ultrasound Beam

IV.1.1 Introduction

As a beam of ultrasound travels outwards from the surface of the transducer, the distribution in space of the ultrasonic energy undergoes change. Axially, the intensity of the beam diminishes gradually with distance along the central axis of the beam, while laterally, at any plane perpendicular to the beam direction, the intensity decreases rapidly with distance from the central axis. Generally, the ultrasound beam spreads out, or undergoes divergence, as it moves away from the transducer. The term "ultrasound beam shape" is commonly used to describe the manner in which the spatial distribution of the beam changes with distance from the source. The beam shape has very significant effects on the quality of the ultrasonic image, and on the tissue depths that can be usefully interrogated using a particular beam. This section examines the factors which influence ultrasound beam shape, and the associated implications for ultrasonic imaging.

IV.1.2 General ultrasound beam shape

The typical manner in which the ultrasound beam spreads out with increasing distance from the transducer, *T*, is shown in Figure IV.1

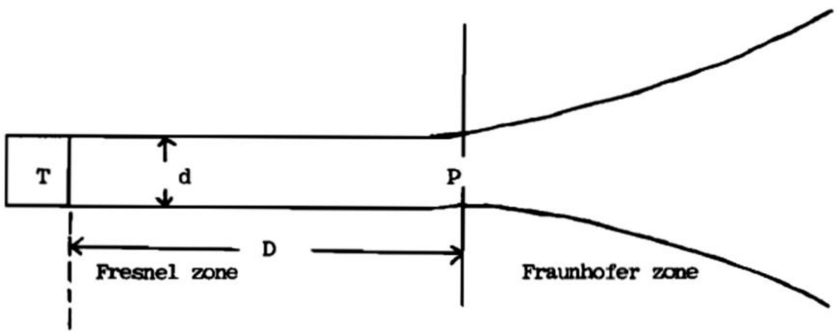


Figure IV.1: General shape of the ultrasound beam.

Initially, between *T* and the plane *P* along the beam path, the beam is narrow, with a small beam width, *d*, equal to about the diameter of the piezoelectric crystal. This part of the beam is referred to as the near field, or the Fresnel zone.

Beyond *P*, the beam spreads out (diverges) over a larger and larger area, with increasing beam widths which result in a rapid deterioration of spatial resolution of the image. This part of the beam is known as the far field, or the Fraunhofer zone.

The distance from the transducer to the plane *P* is sometimes called the transition distance (in reference to the change from Fresnel zone to Fraunhofer zone). The length, *D*, of the Fresnel zone, and the beam width, *d*, at a given plane across the beam, are important parameters which influence, respectively, the practical tissue depth that can be interrogated with the beam, and the spatial resolution in the ultrasonic image.

The narrow beam associated with the near field is desirable for good spatial resolution. The length of this part of the beam therefore determines the approximate tissue depth which, in practice, can be investigated using the beam.

IV.1.3 Factors influencing beam shape

The shape of the ultrasound beam is affected by: the size and shape of the ultrasound source, the beam frequency and the beam focusing

IV.1.3.1 Effect of source size

The size of the ultrasound source affects the beam width, the length of the Fresnel zone, and the angle of divergence beyond the near field. For a transducer in which no focusing is applied, the length, *D*, of the Fresnel zone is determined by the diameter of the transducer and the wavelength of the ultrasound beam according to the relation:

$$D = \frac{r^2}{\lambda} = \frac{d^2}{4\lambda} \quad \ldots\ldots$$

Where r is the transducer radius, λ is the ultrasound beam wavelength and d = 2r is the transducer diameter.

Within the near field, the beam width is approximately equal to the transducer diameter. The length of the Fresnel zone increases rapidly as the beam width (or transducer diameter) is increased.

Conversely, the length of the Fresnel zone diminishes rapidly as the transducer diameter is reduced.

In addition, a small transducer diameter results in a large angle of divergence beyond the near field (see Figure IV.2(a) and Figure IV.2 (b)), thereby diminishing the lateral resolution rapidly.

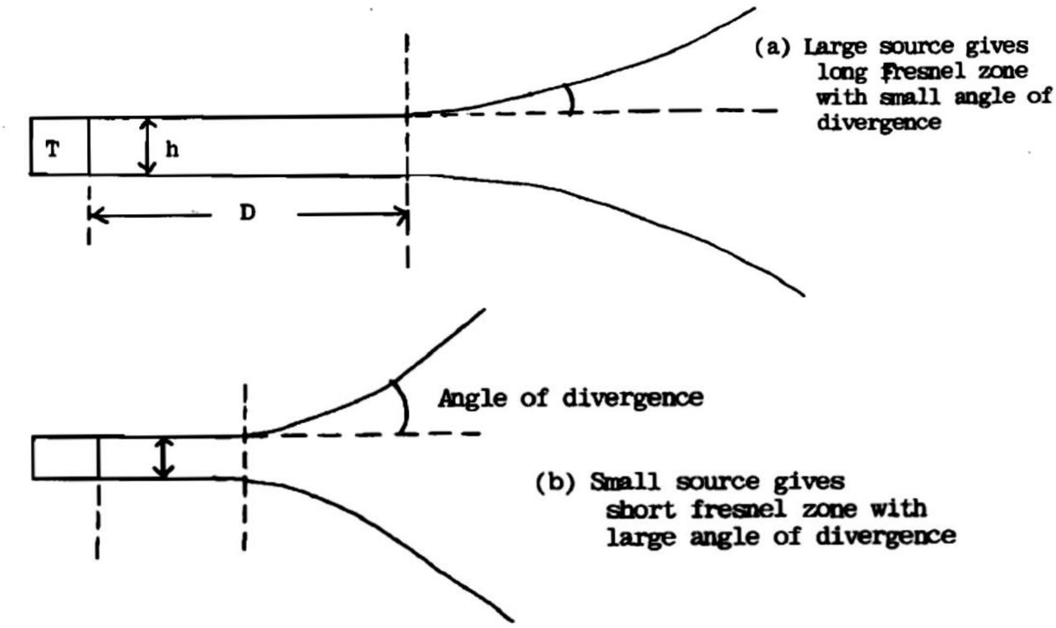


Figure IV.2: Variation of length of Fresnel zone and angle of divergence with source diameter.

In summary, the effects of source size on beam shape are:

- A small source provides a narrow beam initially, is associated with a short Fresnel zone, and the beam diverges rapidly beyond the near field.
- A large source provides a broader beam initially, gives a longer Fresnel zone, and the beam diverges more gradually, thus providing better resolution of deeper structures.

IV.1.3.2 Effect of beam frequency

The above equation can be modified by substituting the wavelength of the ultrasound beam by:

$$\lambda = v/f$$

Then we get

$$D = \frac{d^2}{4\lambda} = \frac{d^2 f}{4v}$$

Where v is the velocity of ultrasound in the transmitting medium, and is the beam frequency.

From this expression, we conclude that the length of the Fresnel zone increases as the beam frequency is increased. Also, the angle of divergence beyond the near field diminishes with increasing frequency. The effect of higher frequencies is therefore not only improved image resolution but also an increase in the length of the useful near field. In practice, however, some of this advantage is taken away by increased beam attenuation at higher frequencies.

IV.1.3.3 Focusing of the ultrasound beam

The shape of the ultrasound beam can be influenced to varying extents by applying different focusing methods.

- ***Shape of the crystal Element***: The crystal element can be suitably shaped by concave curvature to focus the ultrasound beam (Figure IV.3). This is an internal focusing method, because it is effected in the crystal itself. The degree of focusing will depend on the extent of curvature (radius of curvature) of the crystal.

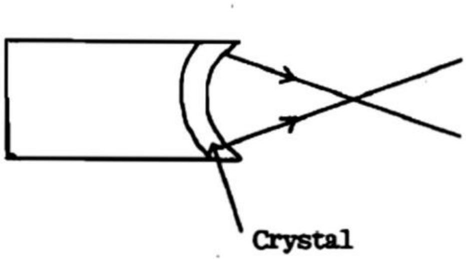


Figure IV.3: Concave shape of crystal.

- ***Acoustic Lenses:*** A concave mirror can be used to focus ultrasound by reflection (Figure IV.4. Again, the degree of focusing will depend on the radius of curvature. Acoustic lenses and mirrors provide external focusing.

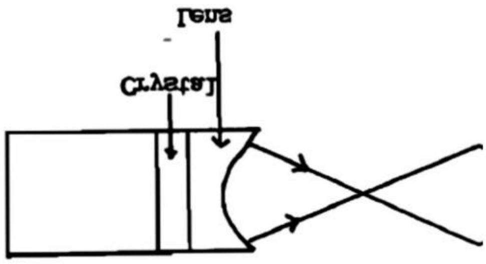


Figure IV.4: Acoustic lens.

- ***Electronic focusing:*** is employed in multicrystal transducers. In such transducers with many crystal elements, movement of the ultrasound beam across the plane of interest in the subject is effected electronically by pulsing small groups of crystal elements at a time. By applying a pulsing programme with carefully controlled time delays between different crystal elements, ultrasound waves from all the crystals in the array can be made to arrive in phase at one particular point (the focus), where they reinforce to produce a high intensity zone. The time delay programme can also be applied during reception of echoes. Electronic focusing offers the advantage of providing variable focus, or dynamic focus, as opposed to the other methods which provide fixed focus. Variable focusing is achieved by altering the time delay programme.
- ***Focus of a transducer, focal zone:*** The focus, *F*, of a transducer is that point along the central axis of the beam which is equidistant in time from all points on the surface of the transducer. The times of flight of the ultrasound waves are equal for all linear paths between *F* and the transducer surface. The waves therefore arrive at *F* in phase and reinforce each other by constructive interference. Attractive beam properties are associated with the point *F*: the beam has its narrowest width, greatest intensity, and best spatial resolution. The focus of a transducer is not sharply defined. Areas within the beam close to *F* will have properties which will closely match those at *F* itself. The region around *F* over which these conditions prevail is called the focal zone of the transducer (Figure IV.5).

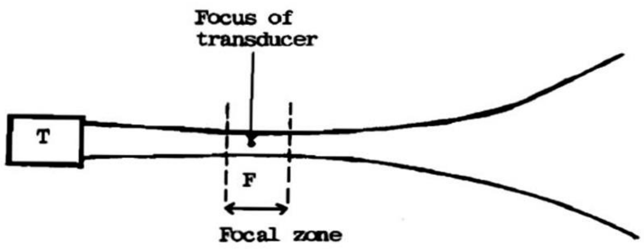


Figure IV.5: Shape of focused beam, showing the focal zone.

The degree of focusing may be classified into three categories as follows: strong focusing (or short focusing), medium focusing and weak focusing (or long focusing). In all cases, fixed focusing gives a focal point which is nearer to the transducer than the transition distance (length of the Fresnel zone). Strong focusing brings the focal point very close to the transducer, typically 2 - 4 cm. It achieves a high degree of beam narrowing, but the beam diverges rapidly beyond the focal distance. It can only be applied to transducers for high resolution examinations of small parts. Weak focusing gives a focal point further away from the transducer - typically more than 8 cm - and a gentle divergence of the beam beyond the focus. It is preferred in diagnostic applications because it provides an extended useful, narrow beam.

IV.2. Echography

IV.2.1 Introduction

Echography is a term made up of two words: ‘’echo’’ and ‘’graphy’’ which means to draw the echo (the reflected wave), which is an imaging technique for visualizing structures of the human and animal body on screen using ultrasound waves. Its history therefore begins with that of ultrasound. It was in 1880 that the brothers Pierre and Jacques Curie discovered the principle of the emission and reception of ultrasound by the phenomenon of piezoelectricity (emission of vibratory waves by a crystal or ceramic subjected to an alternating current and, conversely, creation of an electric flow by a ceramic when it undergoes pressure linked to an ultrasound wave).

The first use of ultrasound was reserved for the army, to detect submarines from the First World War. In 1917, Paul Langevin created the Sonar system which uses the propagation of ultrasound in water. The first use of ultrasound in medicine was made by Dussik, in Austria, in 1947 to explore the brain. The first ultrasound scanner was presented in 1951 in England, designed by Wild, a doctor, and Reid, an electronics engineer. The Doppler effect, discovered in 1942 by Christian Doppler, and allowing the calculation of the radial speed of stars, was then applied to echography and echography in the 1960s for the evaluation of blood flow.

The first indications for echography concerned the brain, then the heart, then the fetus, around the 1970s. Technological progress has allowed for significant developments that continually improve image quality, facilitate the examination and extend the indications to almost all elements of the human body. The most important step was the appearance of real-time probes around the 1980s. Since then, the use of higher frequencies, endocavitary probes and 3D probes around the 2000s have further improved the possibilities of ultrasound investigation in all medical specialties. As equipment manufacturers are always looking for innovations, it is likely that new developments will further extend the applications of echography.

IV.2.2 Echography principle

Echography is a painless and harmless technique used in human and veterinary medicine to study the inside of the human (animal) body, allowing direct observation of internal organs. The echography technique uses ultrasonic waves of frequency ranging from 1 MHz to 20 MHz (up to 50 MHz for the eye), it depends on the organs or biological tissues to be probed. The echo phenomenon is due to the reflection of sound waves on a steep wall, the term graphy designates the written representation of these echoes.

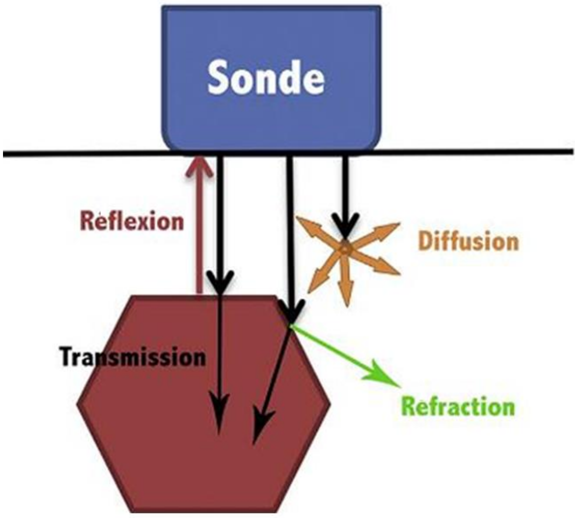


Figure IV.6: Physical phenomena at the origin of the echography image

The echography system uses a probe, a computer system and a visualization system. The element that will emit the ultrasound will be the probe. The latter will in fact send waves in a delimited perimeter. The echography uses different wave frequencies depending on the use that must be made of it.

- 1.5 - 4.5 MHz we can study deep sectors such as the abdomen and this on the order of a few millimeters.
- 5 MHz we will target intermediate structures such as hearts on a scale less than a millimeter.
- 7 MHz we will see small structures close to the skin such as veins or arteries.
- 10 - 18 MHz we will study small animals or we will use it in the context of superficial imaging.

- Up to 50 MHz we will use echography for the observation of the eye.

Before an echography a gel will be applied to the area to be studied to improve the contact between the skin and the probe and to ensure that there is as little interference as possible in the transition of the waves from the probe to the area being studied. Through a probe in contact with the skin, the doctor can view the images obtained on a screen, which allows him to diagnose pathologies without risk or pain for the patient.

To obtain an echography image, the following properties of ultrasound waves are used, among others:

- Speed: The propagation of ultrasound varies according to the media crossed, it is very low in air (340 m/sec), it is done at a speed of approximately 1540 m/sec in soft tissues and water, it is even faster in bone.
- The absorption of the ultrasound wave depends on the frequency of the ultrasound ($a=K.f^2$), the higher the frequency, the greater the absorption phenomenon, which prevents the examination of deep areas at high frequency.
- When it changes medium, part of the incident wave is reflected, the other is transmitted (it continues on its way). We say that there is partial reflection when there is a change of medium at the tissue interfaces.

The echography image is obtained using ultrasound after passing through the different tissues, which are reflected towards the probe. The latter alternately plays the role of transmitter and receiver in extremely short time intervals, of a few fractions of a second.

During the time when the probe receives the ultrasound, the ultrasound scanner analyzes two parameters that influence the image:

- On the one hand, the time taken by the echo to return to the probe, since its emission. This time is double that taken by the ultrasound to reach the interface that created the echo. The speed of propagation in soft tissues and water being constant (1540 m/s), the echography scanner calculates the distance between the probe and the interface to locate the reflection point on the screen, which gives the seat in depth of the interface;

- On the other hand, the intensity of the reflected echo which is proportional to the hardness of the interface that reflected the ultrasound. The boundary between two tissues with very different acoustic impedances (soft tissue/bone or soft tissue/air) gives very intense echoes.

By analogy, we call echo both the wave collected by the probe and the point that represents this echo on the screen. We speak of dense or fine echoes depending on whether they are very strong or weak interfaces. The electrical signal induced by the return echo is amplified and converted, formerly into a cathode current generating an image on an analog screen, now into a transcribed digital signal.

IV.2.3 Formation of the echographic image

The different structures of the human (animal) body give different echographic images depending on their nature:

- Water and liquids (bile, urine, blood, cysts, and effusions) transmit ultrasound perfectly without reflecting it, therefore without giving echoes; they are said to be anechoic, and they appear black on the sections. As they do not cause attenuation of the ultrasound, the organs located further back receive and reflect more ultrasound and appear whiter, this is the "posterior reinforcement" essential to affirm the liquid nature of an image; Ultrasound image of a fluid collection.

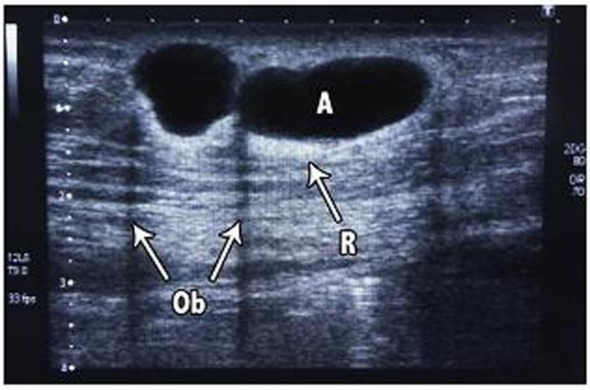


Figure IV.7: Echographic image of a fluid collection.

A: Fluid structure, anechoic (black); ***R:*** Posterior enhancement: ***Ob:*** Edge shadow: artifact due to the refraction of the ultrasound beam which tangentially approaches the contour of the collection and is reflected very obliquely, does not return towards the probe, hence an absence of echoes in the axis of the beam behind the edge of the collection.

- Homogeneous organs with a high water content (liver, spleen, renal cortex, thyroid, salivary glands, prostate, testicles) cause a moderate and regular reflection of ultrasound, they are moderately echogenic and appear medium gray on the sections;
- Heterogeneous tissues and/or composed of fat reflect ultrasound much more, they are hyperechoic and appear very light gray on the sections; on modern screens. Knowing the return times of the echoes, their amplitudes and their speeds, we deduce information on the nature, depth and thickness of the tissues crossed.

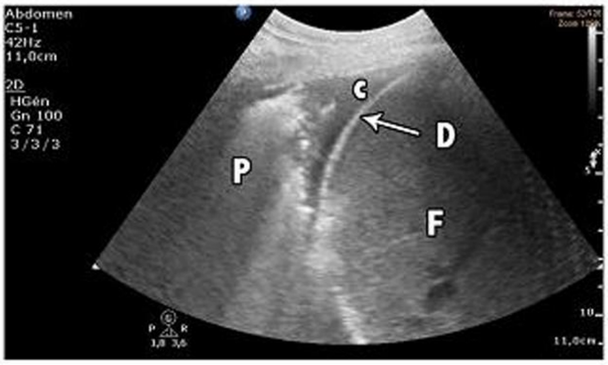


Figure IV.8: Echographic image of the lung-liver interface.

F: Liver. ***D:*** Diaphragm; ***P:*** Lung. Air transmits ultrasound very poorly, the surface of the gas appears hyperechoic but less so than the bone and the shadow cone opposite is less black because the ultrasound diffuses in all directions giving a blurred image; ***C:*** Pleural cul-de-sac, here filled with liquid (pleural effusion), appearing anechoic.

- Muscles and tendons have a fibrillar structure that results in fine echogenic lines; muscles, being much more hydric than tendons, are much more hypoechoic than the latter;
- Bone and calcified elements (stones, vascular or musculoskeletal calcifications) completely reflect ultrasound, their surface appears very hyperechoic, very white and no more echo is detected beyond which forms an "acoustic shadow cone", resulting in a black streak behind the hyperechoic surface.
- Air (pulmonary, intestinal gas) transmits ultrasound very poorly, the surface of the gas appears hyperechoic but less than the bone, and the shadow cone is less black because the ultrasound diffuses in all directions giving a blurred image.

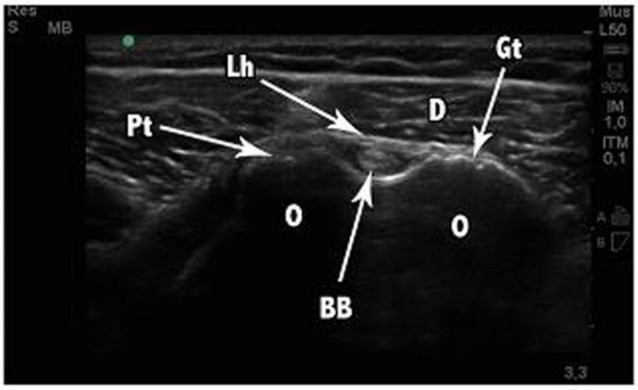


Figure IV.9: Anterior transverse echographic section of the shoulder

Lh: Humeral ligament; ***Pt:*** Lesser trochanter; ***BB:*** Tendon of the long biceps brachi (in its groove); ***D:*** Deltoid muscle; ***Gt:*** Greater trochanter; ***O:*** Acoustic shadow cone (in front of the trochanters of the humerus).

IV.2.4 Different echographic modes

The different echographic modes reflect the tremendous technological improvement of the devices. Currently, all probes are in "real time" and can be used in Time Motion (TM) mode and in Doppler, but it is not useless to recall the other older modes which, being more "basic", allow to understand the elementary mode of formation of the echographic image.

IV.2.4.1 Mode A (Amplitude modulation)

It was the first type of image obtained from the first probes that included a single piezoelectric crystal (single element). It gives an image in a single dimension. It made it possible to search for a displacement of the falx cerebri in intracranial pathologies. The return echo is represented by a peak whose height (on the ordinate axis) is proportional to the intensity of the echo, and whose distance from the emission point (on the abscissa axis) was calculated according to the return time and corresponds to the depth of the interface studied.

IV.2.4.2 Mode B (Brightness modulation)

At the beginning, it was one-dimensional, and represented the echo no longer by a peak but by a point whose brightness was proportional to the intensity of the echo. This additional information on the brightness of the echo made it possible to create the "gray scale". The depth is still calculated as in mode A, but we adopt a more “anatomical” representation, the distance is shown on the ordinate axis in negative values: the interfaces close to the probe appear at the top of the image and, the deeper the interfaces, the more they appear towards the bottom of the ima**ge.**

IV.2.4.3 Mode M (Motion) now TM mode (Time Motion)

These two modes allow to visualize the mobility of a structure as a function of time, in particular for heartbeats. In this mode, the ultrasound emission remains fixed on a constant trajectory and the reflected echoes return to the probe with different delays due to the mobility of the interface, they are represented by points as in B mode. The depth of these echoes therefore varies over time and appears on the ordinate axis, while time scrolls on the abscissa axis.

IV.2.4.4 Two-dimensional imaging in B mode or BD mode

It was a major advance; by making it possible to obtain a "sectional" image in a chosen plane, it allowed the exploration of the abdominal organs. Unlike the M mode where the probe remains fixed and always sends the ultrasounds in the same direction, in this imaging, the probe, always single-element, was mounted on an articulated arm and the sonographer performed a "manual scan", that is to say he moved the probe along a line either longitudinal, transverse or oblique. By summing the echoes collected throughout the scan, a succession of echoes was obtained, the deepest were the lowest in the image because the depth is always on the ordinate axis, those received at the beginning of the slice were on the left of the image and those received at the end of the slice were on the right because the movement of the probe was represented on the abscissa axis.

IV.2.4.5 Real time

This is the examination mode currently generalized on all devices. It allows to have a two-dimensional image with a scan fast enough to also visualize the movement of the organs. It is no longer the sonographer who moves the probe to produce the image, but it is the probes (whose technologies we will see later) which include multiple juxtaposed piezoelectric elements allowing from a single position of the probe to scan an entire plane.

IV.2.4.6 Three-dimensional imaging

It is even more recent, thanks to probes that have even more elements, arranged no longer in a single plane, but over a wider area, allows ultrasounds to be emitted and echoes to be analyzed, no longer in a single plane, but in a volume, which is mainly used for pregnancies.

Figure IV.10: 3D echography.

IV.3. Doppler effect

IV.3.1 Introduction

Doppler echography is a non-invasive two-dimensional ultrasound medical examination that allows to explore intracardiacc and intravascular blood flows. It is based on a physical phenomenon of ultrasound which is the Doppler effect. It is often nicknamed Doppler echography.

- The Doppler effect allows to quantify circulatory speeds.
- Echography allows to visualize vascular structures. In medical practice, Doppler ultrasound is used to explore the arterial network and the venous network in order to evaluate certain conditions: deep vein thrombosis (phlebitis), varicose veins, arteriopathy, thromboses, aneurysms etc.

When two observers are at equal distance from a stationary sound source, they both perceive the same sound. But when the sound source moves toward one observer and away from the other, they each perceive a different sound.

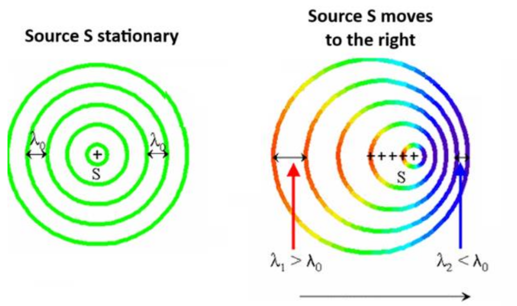


Figure IV.10: Doppler effect principle.

The Doppler effect is a change in the frequency of a wave source when there is relative movement of the source or observer. The frequency perceived by the observer is different from the frequency emitted: it increases if the source or observer gets closer, it decreases otherwise.

The medical application of the Doppler effect was not achieved until the 1960s, with the use of ultrasonic waves.

IV.3.2 Echo-Doppler principle

When an ultrasound beam, emitted by a source, passes through biological tissues, it encounters a number of targets, or fixed interfaces. The frequency reflected by these fixed targets is identical to the emitted frequency: it is said that there is no difference between the emission frequency and the reception frequency. If the target moves, such as the red blood cells in circulating blood, there is a change in the frequency of the reflected beam.

This difference (ΔF) between the emission frequency (F_e) and the reception frequency (F_r) is called the Doppler frequency, it allows the calculation of the speed and direction of the red blood cells and the detection of the movement of the red blood cells in a vessel. The Doppler frequency is expressed by the following relationship:

$$\Delta F = F_r - F_e = \frac{2\,F_e V cos(\theta)}{c}$$

Where F_e is the probe emission frequency; F_r is the probe reception frequency; V is the speed of the elements represented in the vessel; θ is the angle between the axis of the vessel and the axis of the ultrasound beam and c is the average speed of ultrasound in the human body (1,540 m/s).

ΔF is expressed in hertz (Hz), the frequency difference ΔF is positive if the target approaches the source and negative if it moves away from it. In vascular exploration, the value of ΔF is between 50 Hz and 20 KHz which, fortunately, corresponds to a frequency range perceptible to the human ear. F_e is generally between 2 and 10 MHz. The choice of the emission frequency results from a compromise between the attenuation of the ultrasound wave (function of the frequency and depth of the examination) and the backscattering power of the organs which increases with the frequency. The angle θ is a crucial parameter. Indeed,

- For an angle of 90°, $\cos(\theta) = 0$, ΔF is zero, between the vessel and the ultrasound beam, no Doppler signal is obtained.

- For an angle of 0°, $\cos(\theta) = 1$, ΔF is maximum, the ultrasound beam is perfectly in the axis of the vessel, the Doppler effect is maximum.

Calculating the circulatory speed therefore requires knowledge of the Doppler angle. To calculate this speed, the formula becomes:

$$V = \frac{c.\Delta F}{2F_e cos(\theta)_r}$$

IV.3.3 Doppler effect Modes

The Doppler effect can be used in clinical practice in two modes: continuous mode and pulsed mode. Two-dimensional Doppler or color Doppler is based on the principle of pulsed Doppler but the signal processing is different.

IV.3.3.1 Continuous Doppler mode

The probe emits ultrasounds continuously and the frequencies reflected by the red blood cells are analysed continuously, the probe has two piezoelectric crystals, one emitter, and the other receiver. A spectrum of speeds corresponding to all the areas crossed by the beam is thus collected. It allows to record very high velocity flows, without measurable speed limitation, it thus allows to analyse the maximum speed with great precision. Continuous Doppler is very sensitive to detect slow flows. Its disadvantage is a less good localization of the analysed flow.

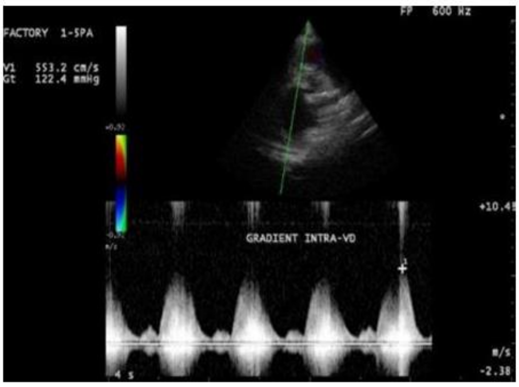


Figure IV.11: Continuous Doppler mode.

IV.3.3.2 Pulsed Doppler mode

It consists of a single crystal probe which alternately emits an ultrasonic beam and (Pulse Repetition Frequency, *PRF*):

$$PRF = \frac{c}{2d}$$

Where c is the ultrasound speed and d is the vessel depth.

The *PRF* determines the depth of the exploration field, because it is necessary to wait for all the echoes to return before emitting a new pulse. The echoes coming from the deepest areas thus set the time interval to be respected before a new shot. The *PRF* also determines the sensitivity to flows. A low *PRF* is necessary to explore in depth and detect slow flows. A high *PRF* is necessary to properly analyse fast flows (avoiding aliasing, which we will come back to). The *PRF* can also be increased if superficial regions are analysed. Between two pulses, the reflected signal is analysed for a very short time called the "listening window". The delay between the end of the pulse and the beginning of the listening window (P) is used to determine the depth of the sampling volume.

The advantage of pulsed Doppler compared to continuous Doppler is that it can benefit from the spatial resolution and focus the examination on a vessel to be analysed. This requires coupling the Doppler analysis with the echographic study. Duplex systems combine the echography image and the Doppler signal. The transmission frequencies are slightly different, the transmission frequency for the Doppler being lower than the frequency of the probe for mode B (for example: 3.5 MHz echographic probe, 3 MHz for the Doppler probe).

The limitations of pulsed Doppler are: its lower sensitivity for detecting very slow flows.

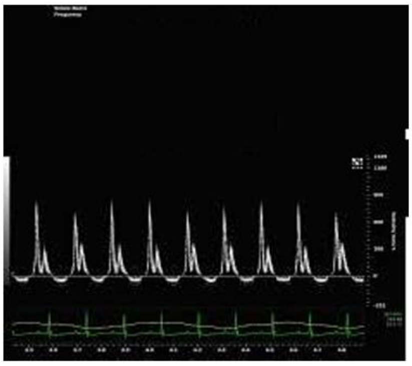


Figure IV.12: Pulsed Doppler mode.

IV.3.3.3 Color Doppler mode

A major technological development then appeared with the integration, in the same equipment, of fast echographic imaging and detection of Doppler information at all points of the echographic image. It could be assimilated to a multi-port multi-line Doppler system, and to differentiate the black and white echographic image from the Doppler signal, the latter was therefore colored by convention. It colors the flows that approach the probe in red and the flows that move away from the probe in blue. The faster the flows, the closer the color is to white.

Color Doppler must face two particular constraints:

- Analyse a very large number of parameters in real time.
- Obtain very quickly the spectral analysis of the signal

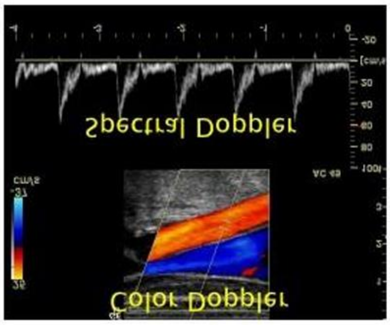


Figure IV.13: Color Doppler mode.

IV.4. Measuring blood flow velocities

IV.4.1 Introduction

Ultrasonic flow meters, which are a type of volumetric flow meter, are non-intrusive flow sensors that use acoustic vibrations to measure the flow rate of a liquid. There are two types, Doppler and transit time. Ultrasonic meters are ideal for automated wastewater applications or any dirty liquid which is conductive or water based - but will generally not work with distilled water or drinking water. These flow meters are also ideal for applications where low pressure drop, chemical compatibility, and low maintenance are all requirements.

In 1842, Christian Doppler discovered that a stationary observer perceives a sound to have shorter wavelengths as its source approaches, and longer wavelengths as its source recedes. This became known as the Doppler effect and explains why one hears rising pitch i n the blowing horn of an approaching car. When the car zooms away, the pitch seems to drop. Ultrasonic Doppler flow meters put this frequency shift to work in so-called dirty liquids containing acoustical discontinuities – or suspended particles, entrained gas bubbles, or turbulence vortexes.

IV.4.2 Ultrasonic flow meter principle

The basic principle of operation employs the frequency shift (Doppler Effect) of an ultrasonic signal when it is reflected by suspended particles or gas bubbles (discontinuities) in motion. This metering technique utilizes the physical phenomenon of a sound wave that changes frequency when it is reflected by moving discontinuities in a flowing liquid. Ultrasonic sound is transmitted into a pipe with flowing liquids, and the discontinuities reflect the ultrasonic wave with a slightly different frequency that is directly proportional to the rate of flow of the liquid. Current technology requires that the liquid contains at least 100 parts per million (PPM) of 100 micron or larger suspended particles or bubbles.

The Doppler ultrasonic flow meter operates on the principle of the Doppler Effect, which is the physical phenomenon of a sound wave changing frequency. In the case of ultrasonic flow meters, the frequency of an ultrasonic signal changes (Doppler Effect) in direct proportion to the rate of flow of the liquid when reflected by suspended particles or gas bubbles (discontinuities) in motion.

Typically, an ultrasonic Doppler flow meter consists of a transmitter/indicator/totalizer and a transducer. The user selects a configuration appropriate to the application – taking into account the liquid, the size and concentration of solids or bubbles, the pipe dimensions and the pipe lining. The transmitter's signal threshold usually adjusts to filter out mechanical and electrical noise. A high-frequency oscillator in the transmitter drives the transducer, which, in the popular clamp-on design, mounts on the pipe exterior. The transducer generates an ultrasonic signal that passes through the pipe wall into the flowing liquid; the transmitter converts the difference between its output and input frequencies to electronic pulses. Processed, scaled, and totalized, the pulses provide a measurement of flow.

Ultrasonic Doppler flow meters that clamp onto the outside of a pipe operate non-invasively, without moving parts. They cause no pressure drop, risk no damage from the process liquid, and entail little maintenance. If properly calibrated, they can have ±1% accuracy – however, the pipe wall and any air space between the wall and the liquid can generate signal interference. Moreover, a stainless-steel pipe wall might conduct the transmitted signal to the extent that the reflected signal will seem to undergo a major shift.

Transit time ultrasonic flow meters measure the time difference between when an ultrasonic signal is transmitted from the first transducer until it is received by the second transducer. A differential comparison is made of upstream and downstream measurements. If there is no flow, the travel time will be the same in both directions. When flow is present, sound moves faster if traveling in the same direction and slower if moving against it.

A third ultrasonic flow meter employs cross-correlation between upstream and downstream transducer pairs to compute flow. Some flow meters of this design use microprocessors to switch automatically between “clean” and “dirty” modes based on correlation factors. A single cross-correlation hybrid flow meter could, for example, monitor flow of either activated or digested sludge. Carefully engineered applications using such flow meters have had reported installed accuracy within 0.5% of reading.

IV.4.3 Ultrasonic flow meter Types

Ultrasonic flow meter technology is a non-contact means of measuring the velocity of a fluid. They are clamp-on devices that attach to the exterior of the pipe (and fit a variety of pipe sizes) and enable measurement of corrosive liquids without damage to the ultrasonic sensor. Portable ultrasonic flow meters are available to aid in industrial application.

The two types of ultrasonic flow meters, Doppler and transit time, each function by way of two different technologies.4

IV.4.3.1 Doppler ultrasonic flow meter

The Doppler ultrasonic flow meter operates on the principle of the Doppler Effect (or Doppler shift), which was documented by Austrian physicist and mathematician Christian Johann Doppler in 1842. He stated that the frequencies of the sound waves received by an observer are dependent upon the motion of the source or observer in relation to the source of the sound. A Doppler ultrasonic flow meter uses a transducer to emit an ultrasonic beam into the stream flowing through the pipe. For the flow meter to operate, there must be particulates, such as solid particles or air bubbles, in the stream to reflect the ultrasonic beam. The motion of particles causes a frequency shift of the beam, which is received by a second transducer. Doppler flow meters are often used for the flow measurement of such fluids as slurries.

Figure IV.14: Doppler ultrasonic flow meter

The meter detects the velocity of the discontinuities, rather than the velocity of the fluid, in calculating the flow rate. The flow velocity (V) can be determined by:

$$V = (f_0 - f_1)\, C_t / 2f_0 \cos(a)$$

Where C_t is the velocity of sound inside the transducer, f_0 is the transmission frequency, f_1 is the reflected frequency, and a is the angle of the transmitter and receiver crystals with respect to the pipe axis. Because $C_t / 2\, f_0 \cos(a)$ is a constant (K), the relationship can be simplified to:

$$V = (f_0 - f_1)\, K$$

Thus, flow velocity V *(ft/sec)* is directly proportional to the change in frequency. The flow (Q in gpm) in a pipe having a certain inside diameter (ID in inches) can be obtained by:

$$Q = 2.45\, V(ID)^2 = 2.45(f_0 - f_1)\,(ID)^2$$

The presence of acoustical discontinuities is essential for the proper operation of the Doppler flow meter. The generally accepted rule of thumb is that for proper signal reflection there be a minimum of 80-100 mg/l of solids with a particle size of +200 mesch (+75 micron). In the case of bubbles, 100-200 mg/l with diameters between +75 and +150 microns is desirable. If either the size or the concentration of the discontinuities changes, the amplitude of the reflected signal will shift, introducing errors.

IV.4.3.2 Transit Time Ultrasonic Flow Meter

Transit time ultrasonic flow meters measure the difference in time from when an ultrasonic signal is transmitted from the first transducer until it crosses the pipe and is received by the second transducer. A comparison is made of upstream and downstream measurements. If there is no flow, the travel time will be the same in both directions. When flow is present, sound moves faster if traveling in the same direction and slower if moving against it. Since the ultrasonic signal must traverse the pipe to be received by the sensor, the liquid cannot be comprised of a significant amount of solids or bubbles, or the high frequency sound will be abated and too weak to travel across the pipe.

Figure IV.15: Transit Time Ultrasonic Flow Meter.

When the flow is zero, the time for the signal T_1 to get to T_2 is the same as that required to get from T_2 to T1. When there is flow, the effect is to boost the speed of the signal in the downstream direction, while decreasing it in the upstream direction. The flowing velocity (V_f) can be determined by the following equation:

$$\boldsymbol{V_f = K\, dt/T_L}$$

Where K is a calibration factor for the volume and time units used, dt is the time differential between upstream and downstream transit times, and T_L is the zero-flow transit time.

Theoretically, transit-time ultrasonic meters can be very accurate (inaccuracy of ±0.1% of reading is sometimes claimed). Yet the error in these measurements is limited by both the ability of the signal processing electronics to determine the transit time and by the degree to which the sonic velocity (C) is constant. The speed of sound in the fluid is a function of both density and temperature. Therefore, both have to be compensated for. In addition, the change in sonic velocity can change the refraction angle (“a” in Figure 1B), which in turn will affect the distance the signal has to travel. In extreme cases, the signal might completely miss the downstream receiver. This type of failure is known as walk-away.

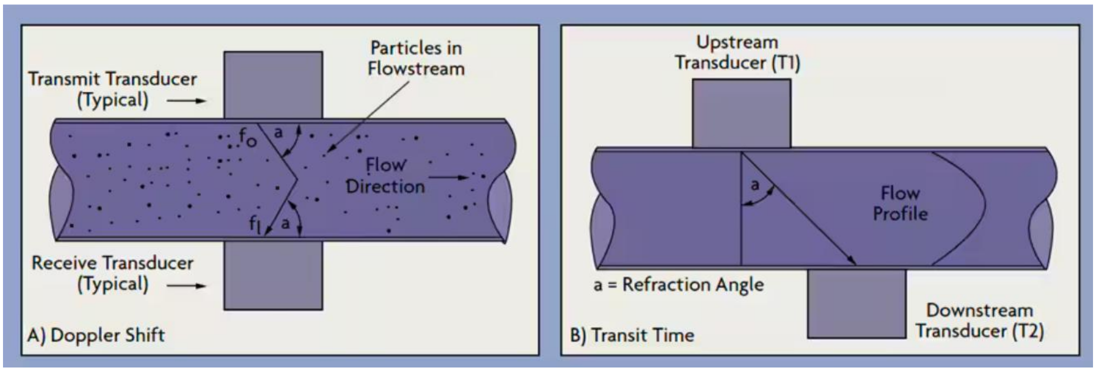


Figure IV.16: Flow Meter Designs.

IV.5. Bone Density Measurement

IV.5.1 Introduction

Bone is unique in its ability to maintain its strength in the presence of constant turnover. It has been established that there is gradual and relatively complete removal and replacement with new bone several times in a lifetime. The reparative pattern is constantly changing in order to maintain strength and is influenced by changes in general health, the presence of disease, and the degree of physical activity. To date, no method has been developed to measure the nature of lifetime changes in patterns during childhood and throughout older age. Often the first indication that a problem exists is the occurrence of a fragility fracture, which is often associated with a high degree of morbidity and mortality.

In an attempt to address many of the above issues, ultrasound has been proposed. It was first described for clinical bone assessment in 1984 by Langton. The great promise of ultrasound is related, in part, to the fact that as a mechanical wave it is affected by the biomechanical and biophysical properties of a bone through which it propagates. Prior research and recent observations have enlarged the scope of ultrasound's capability for detecting changes in bone and measuring a variety of features, including architectural structure of bone and mass. In view of these and other emerging observations to be mentioned regarding the application of ultrasound to assess bone quality in diagnosis, screening, and monitoring the course of osteoporosis.

IV.5.2 Ultrasonic Bone Density Analyser principle

Bone Density Measurement (BMD) has been generally measured using dual energy X-ray absorptiometry (DEXA) by comparing the image density of the skeleton to the surrounding soft tissues. The mineral density of bone is estimated by the degree of grey colour of the X-ray image in the bone region. However, DEXA has not become a standard tool in the clinic of a primary care physician as it is expensive and inconvenient and involves X-ray exposure. Another commonly used method for BMD measurement is based on the use of ultrasound as an alternative to DEXA. Quantitative ultrasound (QUS) technique has become quite popular in recent years due to its several advantages that it is non-ionizing, relatively less expensive, and simple to use. Ultrasound employs high frequency sound waves for measurement of BMD. The ultrasound waves are produced by piezoelectric crystals mounted in the transmitting and receiving transducers. The measurement is usually done on the heel, shin, or finger. The process is quite simple and involves asking the person to place the bare foot in the device as shown in Figure IV.17.

Most devices choose the heel as the measurement site because it has a large volume of trabecular bone between relatively flat faces and is readily accessible for transmission measurements. The transducers are acoustically coupled to the heel by elastomer transducer pads using ultrasound coupling gel, which is applied to the transducer pads. This ensures a good acoustical contact with the skin over the bone being tested. The heel is positioned between a pair of ultrasound transducers, with one transducer transmitting the ultrasound signal and the other transducer receiving the signal after passage through the heel. The inbuilt computer calculates the bone density by measuring as to how fast the sound waves pass through the heel. The measurement and generation of the report just takes less than a minute.

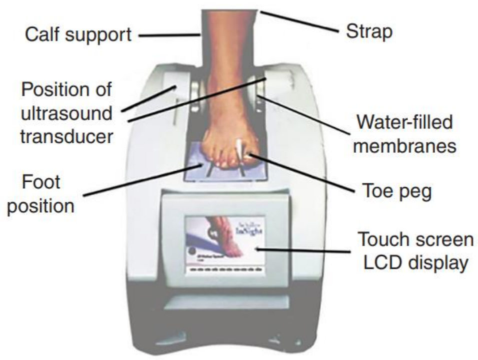


Figure IV.17: Ultrasound bone density analyser.

The ultrasound frequency applied in QUS generally range between 0.2 and 1.5MHz. There are two types of QUS techniques in use depending on the axis the ultrasound waves can take to travel through the bone. These can be either horizontal or longitudinal. Horizontal transmission uses probes that measure the speed of sound (SOS) on the cortical layer of the bone at a fixed distance, whereas the longitudinal transmission is more often used in clinical practice and is mainly applied to the heel bone (calcaneus). Calcaneus is considered the most reliably reachable location due to the presence of a large semi-flat region of almost exclusively cancellous bone, which can give accurate and reliable results.

IV.5.3 Ultrasonic Bone Density Analyser Types

From the signal measured by the receiving transducer, two parameters describing the nature of the received sound waves can be simultaneously determined: Speed of Sound and broadband ultrasonic attenuation. It has been observed that broadband ultrasonic attenuation is reduced in patients with osteoporosis, because there are fewer trabecular in the calcaneus to attenuate the signal. Similarly, Speed of Sound values are reduced in patients with osteoporosis because, with the loss of mineralized bone, the elastic modulus of the bone is decreased.

IV.5.3.1 Speed of Sound Method

During passage through the bone, the speed and amplitude of the ultrasound wave gets attenuated with the changes detected by the receiving transducer. Since the Speed of Sound in bone tissue is higher than in soft tissue, this information is directly related to the amount of bone in the ultrasound path. Ultrasound speeds of 2800–3000m/s are typical in cortical bone, whereas speeds of 1550–2300m/s are typical in trabecular bone. In addition, time differences between the arrival times of the first arriving signal and a more energetic guided wave component can be used to gauge the amount of bone. Many commercial QUS devices measure the Speed of Sound through the heel.

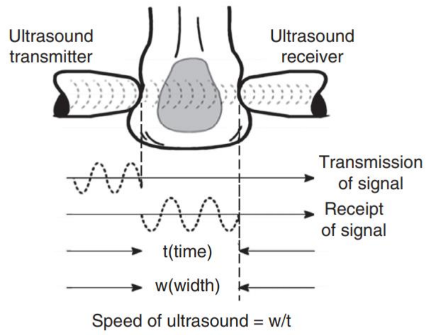


Figure IV.18: Principle of measurement of speed of sound for bone density measurement.

Speed of Sound is determined by measuring the width of the heel, and the time delay between the initial transmission of the sound waves from the transmitting transducer and the receipt of the sound waves by the receiving transducer as shown in Figure IV.18.

The system measures the time (*t*) the ultrasound signal takes to go through the heel and the transducer pads minus the propagation time measured with the pads touching each other and with no heel interposed. The width of the heel (*w*) is measured using a micrometre attached to the transducers. The Speed of Sound value is then equal to w/t and is measured in metres per second (m/s). The range observed with in a typical population is approximately 1450–1700m/s, with young/ healthy subjects having higher Speed of Sound values than older or osteoporotic subjects.

IV.5.3.2 Broadband Ultrasound Attenuation (BUA) Method

An ultrasound pulse passing through bone is strongly attenuated as the signal is scattered and absorbed by trabeculae. Attenuation is measured in decibels. The rate of ultrasound attenuation, or loss of energy, over a given frequency range is referred to as the rate of 'broadband ultrasound attenuation'. Its value increases linearly with frequency, and it is usually expressed as decibels per megahertz (dB/MHz). The slopes of the frequency spectrum reflect the density and structure of bone. It has been observed that the SOS is reduced and broadband ultrasound attenuation (BUA) is decreased with reductions in bone density and trabecular number, correlating well with the physical density of the bone under study leading to a diagnosis of osteopenia or osteoporosis. The range of BUA observed in a typical population is approximately 30–130dB/MHz, with young/healthy subjects having higher BUA results than older or osteoporotic subjects.

Attenuation of the ultrasound pulse depends upon a number of factors such as scattering, diffusion, absorption, transmission, reflection, etc., and therefore has a complex relation with the amount and material properties of bone. Also, attenuation shows a linear relationship to frequency in the 0.2–0.6MHz range, with the bone attenuating high frequency sound waves much more than low frequency sound waves. The slope of the attenuation curve with respect to frequency has been shown to depend on bone mineral content (BMC) and BMD.

Just like DEXA, bone density measurements are reported as a value in g/cm2 and as a T-score or Z-score, which reflects the bone density measurement in relation to other people in a similar group, termed as reference population. The reference population for the T-score is young adults of the same sex as the patient, while the reference population for the Z-score is a group of the same age and sex as the person being tested.

Commercial machines for quantitative real-time bone assessment are usually handheld and portable and can operate on rechargeable batteries. These devices are constructed around microprocessors or a digital signal processor, using high resolution A/D converter that samples the ultrasound signal at 30 MHz. In one design, the pulser produces a 300V 400 ns pulse that excites the 2.25 MHz 12.7 mm diameter transmitting transducer. The 12.7 mm coaxially located receiver transducer operates at 1 MHz that receives the ultrasound signal after it has propagated through the heel of a foot. Both the transducers are flat and unfocused. The combination of the higher frequency transmitter with the lower frequency receiver allows to increase the bandwidth of the overall system. A high resolution electronic calliper (0.03 mm) is included in the device that measures the distance between the transducers. The adjustment of gel pads is automatic for optimal foot positioning. The device measures the SOS and BUA, thus allowing to calculate the bone quality index (BQI).

With the developments in QUS, several new diagnostic parameters have emerged from experimental results conducted both in vitro and in vivo. These parameters have shown potential for assessing not only bone quantity but also bone quality in terms of structure and strength.

Chapter-V:

Sound Waves in Prospecting & Industry

V.1 Seismic prospecting

V.1.1 Introduction

Seismic prospecting is based on the propagation of elastic waves in the subsoil. We indicated in the previous section that we were mainly interested in volume waves, namely expansion-compression waves (P waves) and shear waves (S waves).

You saw during our field trip how seismic prospecting is implemented. We cause an impact with a source (e.g. a rifle or a sledgehammer blow) and the seismic waves that are generated by the source propagate in the subsoil. When they arrive at an interface between two layers of different speeds, part of the waves are reflected towards the surface, the other part being transmitted in the deeper layers. The angles of incidence, reflection and transmission are linked by the ray parameter, *p*, which is constant for each ray

$$p = \frac{\sin\theta_i}{V_i} = \frac{\sin\theta_r}{V_r} = \frac{\sin\theta_t}{V_t}$$

Where $\theta_{i,\,r,\,t}$ and $V_{i,\,r,\,t}$ are the angles and velocities in the incident, reflected and refracted media. We will focus here first on the reflected waves, then on the transmitted waves.

V.1.2 Seismic reflection

Let us consider a seismic wave emanating from a source S and incident on an interface between two media of constant speeds V_1 and V_2. The layer of speed V_1 has a thickness h. The reflected wave is recorded by a receiver (i.e. a geophone or a hydrophone) at a distance x from the source (Figure V.1).

We will solve this problem based on ray theory, which has the advantage of reducing these problems to simple questions of geometry. Note that rays have no physical existence: it is not the passage of a ray that is recorded with a geophone, but that of a wave front. We can of course solve these problems by using wave fronts, but the formalism is more complicated.

We will try to calculate the round trip time between the source and the receiver. To do this, since the speed is constant in the layer, we only have to take

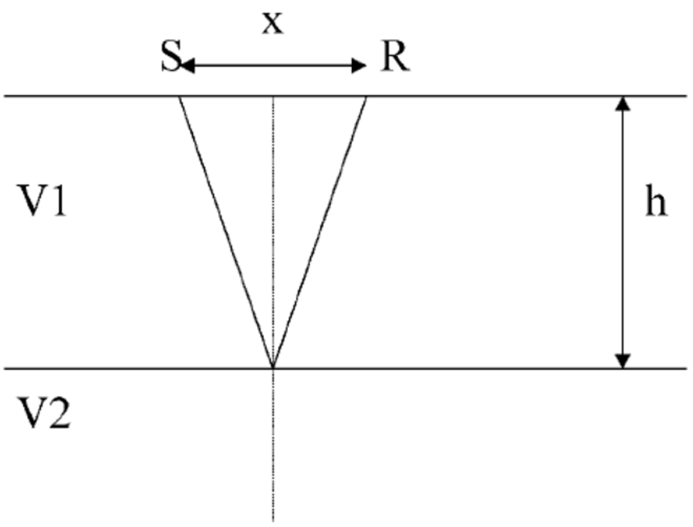


Figure V.1: Geometry for the seismic reflection problem.

The distance travelled divided by the speed. Following Figure V.1, we find that this distance is given by:

$$d = 2\sqrt{\left(\frac{x}{2}\right)^2 + h^2} = \sqrt{x^2 + 4h^2}$$

So, the travel time is:

$$t = \frac{1}{V_1}\sqrt{x^2 + 4h^2}$$

$$t^2 = \left(\frac{x}{V_1}\right)^2 + \left(\frac{2h}{V_1}\right)^2$$

This curve describing the relationship between the travel time and the source-receiver distance is known as the hodochrone. The characteristic hodochrone of a reflected wave is a hyperbola, i.e. if you see a hyperbola on a seismic shot, you are dealing with a reflected wave.

Note that only the first term depends on the distance *x*. The second term only depends on the parameters of the layer: it is in fact the round trip time between the surface and the bottom of the layer defined by $t_0 = 2h/V_1$. Our hodochrone therefore becomes:

$$t^2 = \left(\frac{x}{V_1}\right)^2 + t_0^2$$

Presence of a sloping layer? The geometry of the problem is shown in Figure v.2. Note that in this case the reflection point (Q) is not exactly halfway between the source and the receiver.

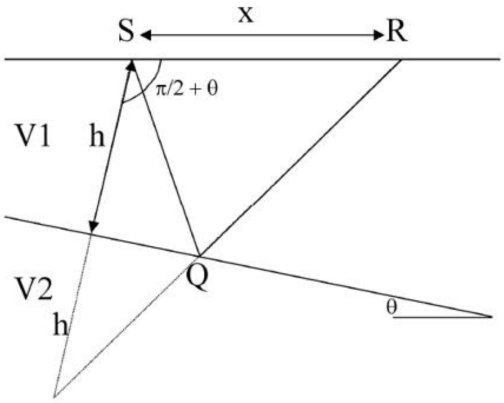


Figure V.2: Geometry for the problem of seismic reflection for a layer inclined at an angle ''*θ*''.

To solve this problem, we proceed as in the previous case, i.e. we determine the distance travelled and divide it by the speed of the medium. To make things easier, we rotate the path *SQ* relative to the reflecting plane: the total path is then from the intersection of the two dotted lines (*P*) to point R.

We find this distance *PR* by applying the Law of Cosines, because the triangle *SPR* does not contain any right angles.

$$(PR)^2 = d^2 = x^2 + (2h)^2 - 4hx\cos(\pi/2 + \theta)$$
$$d^2 = x^2 + 4h^2 + 4hx\sin\theta$$
$$d^2 = x^2 + 4hx\sin\theta + 4h^2\sin^2\theta + 4h^2\cos^2\theta$$
$$d^2 = (x + 2h\sin\theta)^2 + (2h\cos\theta)^2$$

Where we get the travel time from

$$t^2 = \left(\frac{x + 2h\sin\theta}{V_1}\right)^2 + \left(\frac{2h\cos\theta}{V_1}\right)^2$$

We see that the hodochrone still describes a hyperbola. Let's look at the second term: as in the previous case, it does not depend on *x*. We see that in fact this term is none other than the t_0 of the previous case divided by $\cos\theta$; in other words, it is the t_0 for a propagation speed $V_v = V_1/\cos\theta$. V_v is the apparent vertical speed higher than V_1.

The analysis of a hodochrone therefore gives us information on the speed of the layer (which controls its curvature) and on its thickness. But we saw during the field camp that the seismic signals were of more or less large amplitude. Can we use this to find out a little more?

The amplitude of a reflected wave depends essentially on the impedance contrast between the two media. The impedance is defined as the product of the velocity and the density $Z = \rho V$. The reflection coefficient R, for a wave incident normally on an interface is given by:

$$R = \frac{Z_2 - Z_1}{Z_2 + Z_1} = \frac{\rho_2 V_2 - \rho_1 V_1}{\rho_2 V_2 + \rho_1 V_1}$$

So, the greater the difference between two media, the greater R will be. Also note that R is positive from a slow medium to a fast medium and vice versa.

V.1.3 Seismic refraction

Now let's look at what happens in the velocity layer V_2. We saw above that the ray parameter p is constant. Therefore, the trajectory of the ray in medium 2 is given by:

$$\sin\theta_t = \frac{V_t \sin\theta_i}{V_i}$$

We therefore see that if $V_t > V_i$, the ray will be further from the normal in medium 2 than in medium 1. We can even imagine a case for which $\sin\theta_t = 1$, or $\theta_t = \pi/2$. This case will occur when

$$\sin\theta_i = \frac{V_i}{V_t} = \sin\theta_c$$

Where θ_c is the critical angle, that is, the angle at which the ray propagates along the interface in medium 2. We speak of a critically refracted wave or a conical wave.

We will calculate the hodochrone for a cone wave propagating along a plane interface (Figure V.3). Our approach will be "divide and conquer", i.e. we will look at the paths in layers 1 and 2 separately.

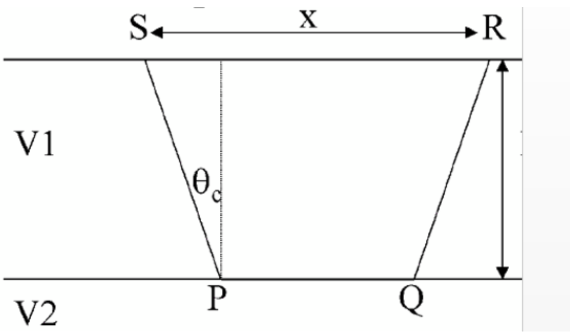


Figure V.3: Geometry for the problem of seismic refraction

In layer 1, the path is:

$$d = SP + QR$$

$$d = \frac{2h}{\cos\theta_c}$$

So, the travel time is:

$$t_1 = \frac{d}{V_1} = \frac{2h}{V_1 \cos\theta_c}$$

In layer 2, the path is:

$$d = PQ$$

$$d = x - 2h\tan\theta_c$$

$$t_2 = \frac{x - 2h\tan\theta_c}{V_2}$$

Let's combine the two times to find the total time *t*

$$t = t_1 + t_2$$

$$t = \frac{2h}{V_1\cos\theta_c} + \frac{x - 2h\tan\theta_c}{V_2}$$

$$t = \frac{x}{V_2} + \frac{2h}{V_1\cos\theta_c}\left(1 - \frac{V_1\sin\theta_c}{V_2}\right)$$

But $V_1/V_2 = sin\ \theta c_\mu$, hence:

$$t = \frac{x}{V_2} + \frac{2h\cos\theta_c}{V_1}$$

$$t = \frac{x}{V_2} + \frac{2h}{V_1}\sqrt{1-\left(\frac{V_1}{V_2}\right)^2}$$

We see that here, the hodochrone is a simple straight line with slope 1/V2 and whose ordinate at the origin depends only on the thickness of the layer and the speeds on either side of the interface.

We can generalize this relationship for a medium with *N* speeds (we will not demonstrate this here):

$$t_N = \frac{x}{V_N} + \sum_{i=1}^{N-1}\frac{2h_i}{V_i}\sqrt{1-\left(\frac{V_i}{V_N}\right)^2}$$

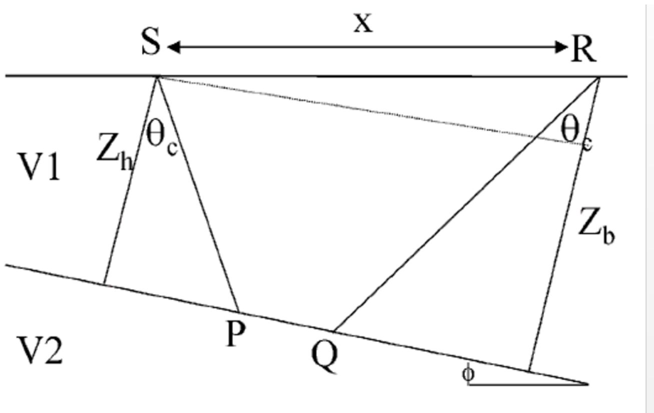


Figure V.4: Geometry for the problem of seismic refraction;

Where t_N is the cone wave propagating at the interface above layer *N*, i.e. the deepest interface. If we are interested in a shallower interface, we only have to choose an *N* immediately greater than the desired interface.

As in the case of reflection, it is desirable to calculate the hodochrone for a steep interface. The geometry of this problem is given in Figure V.4.

As in the previous case, let us divide and conquer. The total travel time is the sum of the times in each medium. According to Figure V.4,

$$t = \frac{SP + QR}{V_1} + \frac{PQ}{V_2}$$

$$SP = \frac{Z_h}{\cos\theta_c}$$

$$QR = \frac{Z_b}{\cos\theta_c}$$

$$PQ = x\cos\phi - (Z_h + Z_b)\tan\theta_c$$

$$t = \frac{Z_h + Z_b}{V_1\cos\theta_c} + \frac{x\cos\phi}{V_2} - \frac{(Z_h + Z_b)\tan\theta_c}{V_2}$$

$$t = \frac{x\cos\phi}{V_2} + \frac{Z_h + Z_b}{V_1\cos\theta_c}\left(1 - \frac{V_1\sin\theta_c}{V_2}\right)$$

But $V_1/V_2 = \sin\theta c_\mu$, hence:

$$t = \frac{x\cos\phi}{V_2} + \frac{Z_h + Z_b}{V_1\cos\theta_c}\left(1 - \sin^2\theta_c\right)$$

$$t = \frac{x\cos\phi}{V_2} + \frac{Z_h + Z_b}{V_1}\cos\theta_c$$

We are therefore faced with two cases: either the shot is fired from *S* to *R* (direct shot) or from *R* to *S* (reverse shot). Let us determine the hodochrones for these two cases.

1. Shot in S. We then shoot down the slope. As we know that Zh and Zb are connected by x and φ, we will eliminate Z_b using the relation $Z_b = Z_H + x \sin\varphi$

$$t_{SR} = \frac{x\cos\phi}{V_2} + \frac{2Z_h}{V_1}\cos\theta_c + \frac{x\sin\phi}{V_1}\cos\theta_c$$

But sin $\theta c = V_1/V_2$ therefore:

$$\frac{x\cos\phi}{V_2} = \frac{x\sin\theta_c\cos\phi}{V_1}$$

Let's substitute

$$t_{SR} = \frac{x}{V_1}(\sin\theta_c \cos\phi + \cos\theta_c \sin\phi) + \frac{2Z_h}{V_1}\cos\theta_c$$
$$t_{SR} = \frac{x}{V_1}\sin(\theta_c + \phi) + \frac{2Z_h}{V_1}\cos\theta_c$$

2. Shooting in *R*. We then shoot up the slope. We eliminate Z_h using $Z_h = Z_b - x \sin \varphi$. Following the same reasoning, we find:

$$t_{RS} = \frac{x\cos\phi}{V_2} + \frac{2Z_b}{V_1}\cos\theta_c - \frac{x\sin\phi}{V_1}\cos\theta_c$$
$$t_{RS} = \frac{x}{V_1}(\sin\theta_c \cos\phi - \cos\theta_c \sin\phi) + \frac{2Z_b}{V_1}\cos\theta_c$$
$$t_{RS} = \frac{x}{V_1}\sin(\theta_c - \phi) + \frac{2Z_b}{V_1}\cos\theta_c$$

In both cases, we notice that the hodochrone is a straight line, but the slopes (i.e. the speeds) are different.

$$V_{SR} = \frac{V_1}{\sin(\theta_c + \phi)} \qquad V_{RS} = \frac{V_1}{\sin(\theta_c - \phi)}$$

These apparent speeds allow us to determine, without any calculation, where the top and bottom of the interface slope are. Indeed, since θc and φ are positive numbers, we know that $\theta c + \varphi > \theta c - \varphi$ and therefore that $VSR < VRS$. The shot with the highest apparent speed is above the bottom of the slope. But we can go further and determine the two angles θc and φ from these apparent speeds.

$$\theta_c = \frac{1}{2}\left[\arcsin\left(\frac{V_1}{V_{SR}}\right) + \arcsin\left(\frac{V_1}{V_{RS}}\right)\right]$$
$$\phi = \frac{1}{2}\left[\arcsin\left(\frac{V_1}{V_{SR}}\right) - \arcsin\left(\frac{V_1}{V_{RS}}\right)\right]$$

From which we the6n draw V_2, Z_h and Z_b.

We therefore see the importance of carrying out at least one direct shot and one reverse shot in order to determine the dip. If we only have one shot at our disposal, we are obliged to interpret our seismic-refraction data in terms of flat layers.

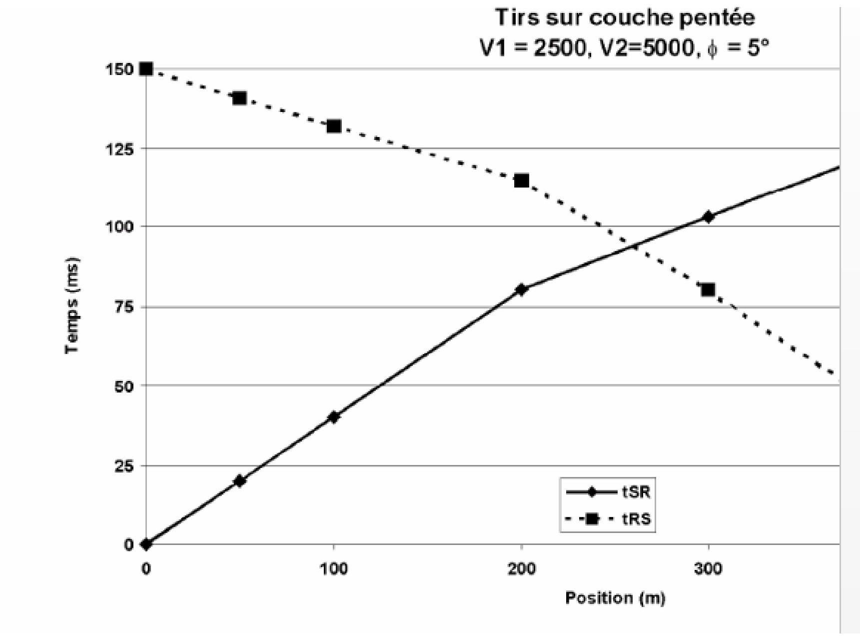


Figure V.5: Example of direct and reverse firing hodochrones on a sloping layer.

V.2 Underwater detection

V.2.1 Introduction

The ocean is the beating blue heart of the planet and the largest habitat on Earth. It covers approximately 70% of the Earth's surface and holds 97% of the Earth's water. It plays a significant foundational role in supporting both the environment and the economy. The environment produces more than half of the oxygen and absorbs most of the carbon. It also provides the biggest habitats for natural organisms. Sea grass meadows and coral reefs are the foundation of marine food chains, habitat provision, and nutrient cycling. If marine species (such as coral reefs and sea grasses) are damaged, the global climate will have largely deteriorated and human food resources will be largely affected. For the economy, maintaining the ocean ecosystem services is also important. Scientific ocean exploration and exploitation help to effectively manage, conserve, regulate, and use ocean resources. However, the understanding of the ocean remained limited for a long time due to scarce tools and technologies to explore the ocean world. Fortunately, the underwater detection technique, as an effective way to explore the ocean, offers enormous potential to make exploration and exploitation more intelligent and efficient. In the past, human beings explored and exploited the ocean traditionally. For example, in the ocean research field, researchers try to quantify human impacts on fish biodiversity to preserve marine ecosystems. People fish by fishnet or visit the fish community by diving into the ocean world. These quantification approaches help us to conduct in situ sampling of the fish community, but cannot collect sufficient data and require considerable human physical resources. In recent years, Autonomous Underwater Vehicles (AUVs) and Remotely Operated Vehicles (ROVs) equipped with intelligent underwater object detection systems have offered us opportunities to explore and protect ocean resources.

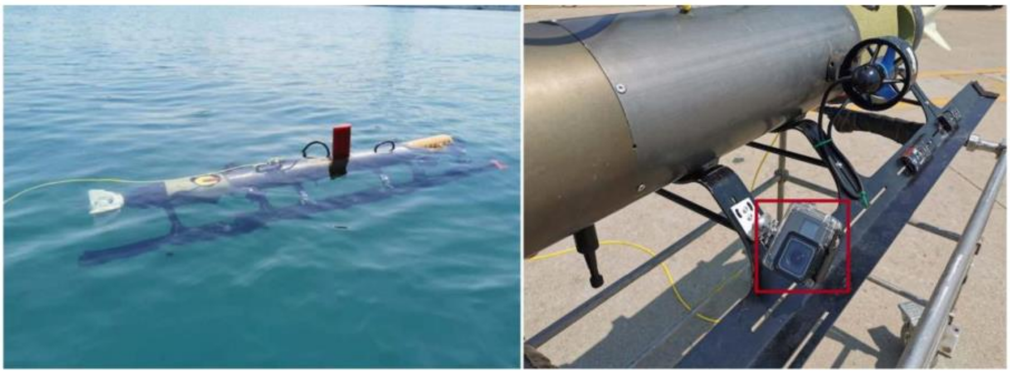


Figure V.6: Autonomous Underwater Vehicles (AUVs) Equipped with GoPro Cameras.

Underwater cameras on AUVs and ROVs assist in monitoring and performing underwater tasks, such as capturing, surveilling, and counting marine organisms. These tasks require intelligent object detection systems, which face challenges from noisy underwater images and videos. As shown in Figure V.6, researchers used an AUV with a camera system to collect and analyze underwater data. Underwater vision-based applications are useful for marine ecology and management. However, robust object detection techniques are needed to overcome the underwater challenges for AUVs and ROVs.

V.2.2 Challenges of Underwater Object Detection

The underwater environment is one of the most challenging conditions for object detection. Underwater object detection faces several challenges:

1. Underwater images collected in the underwater scenes are of very low quality and contain considerable small-size objects degrading the accuracy of the detection frameworks. In underwater scenes, the light received by any camera suffers from wavelength-dependent light absorption and scattering caused by the particles in the water. Light absorption leads to the loss of image information and serious color distortion while

light scattering produces haze-effects, reduces image contrast, and suppresses image details. Moreover, the detection performance has also been affected by shadow noise, non-uniform illumination, camera shaking, and complex background interference. Hence, underwater image enhancement, as the pre-processing procedure, is commonly used to assist underwater vision tasks.

2. Well-annotated underwater data is not sufficient in terms of diversity and amount. The performance of machine learning models is highly influenced by the size of the available datasets. For example, deep learning is a part of machine learning, and the established deep learning models trained on few training examples usually present poor performance, because they are likely to suffer from the over-fitting problem (a model fits exactly on the training data but cannot perform accurately on new data). For the underwater detection problem, previous detection frameworks have been designed and evaluated on underwater datasets collected under constrained conditions. The scale of the datasets is very small because the annotation of the underwater objects is extremely labor- and time-consuming. As reported in, most of the constructed underwater data has little human annotations.

3. Current state-of-the-art detection frameworks have limited generalization and robustness capabilities. They cannot learn effective feature representation of datasets with imbalanced data distributions and imbalanced label noise distributions. Large-scale datasets with balanced and correct human annotations bring large performance advantages to machine learning models, however, machine learning models may fail when the annotation quality of the training data cannot be guaranteed

V.2.3 Characteristics of Underwater Images

In contrast to traditional photography capturing the expanse of open skies, underwater imagery is imbued with a vivid prevalence of blue and green hues. However, the formidable light attenuation within the aquatic medium, unlike the open air, along with a heightened diffusion of incident light, yields a consequential reduction in perceptibility. As a result, objects located at significant distances from both the observer and the imaging apparatus, as well as those at intermediate distances, and sometimes even those situated closer, are hardly visible and display minimal contrast when compared to their immediate environment. Furthermore, the presence of suspended particulates within the water, encompassing elements such as sand, plankton, and algae, engenders a phenomenon where incident light interplays with these particles, bestowing upon them reflections and distinct forms.

V.2.4 Underwater Image Processing

In the field of underwater image processing, two crucial factors come into play: the inherent properties of the water medium and the complex behavior of light as it traverses water. Understanding these factors is essential for effective underwater image processing, regardless of the application. Unlike air, water has distinct properties that lead to light degradation effects in captured images, setting them apart from airborne images. This divergence affects the visual scene by reducing contrast, creating haze, and attenuating light. Light attenuation, in particular, is a primary cause of the haziness in underwater images. As light travels through water, it gradually dissipates and fades, typically occurring at around 20 meters in clear water. In contrast, in turbid water, light diminishes after traveling a short distance, often less than 5 meters.

- Due to light attenuation, in clear water, the visibility goes off after 20m at about twenty meters and in turbid water, the visibility vanishes at about 5 meter distance. Absorption (defusing the light energy) and scattering (alters the light reflection path) are major cases of the light attenuation process.
- The scattering and absorption effects due to the light attenuation process, make the underwater image quality lower, and underwater image visual quality is suffered in a water environment. Light deviation from its path from an object to the camera is called forward scattering and generates a blurring effect on an image. In contrast, some part of light reflection from the middle without reaching the object is called backward scattering generally limits the contrast of the images. It results in inconsistent lighting, reduced clarity, weaker contrast, and color shifts towards green and blue hues. Light deviates when encountering objects, causing forward scattering and blurriness. Backward scattering occurs when light reflects without hitting an object, reducing image contrast.
- The light intensity gets lower as light travels in water and the light color components gradually disappear depending upon their wavelengths. Due to the shortest wavelength, the blue color penetrated underwater more as compared to other components of the light, making the underwater images bluish due to blue color light component dominance.

- The underwater images suffer from poor quality and suffer from these problems: un-even lighting, poor visibility, low contrast, blurring, disappearing of color components (resulting in a greenish and bluish appearance), and noise. Due to these problems, any computer vision application targeting underwater images needs to rectify one or more issues from these.

There are two different strategies to process an image. one is image restoration and the other is image enhancement. The details are: 1. In image restoration, a degradation model is suggested to recover the poorly constructed image quality. The original image is required to recover the quality of an image. These methods are reliable but they need many environment property values as parameters for their model. These parameters can be light attenuation and water turbidity diffusion coefficients. These parameters vary from time to time and have variable values. And underwater depth of the scene is also an important parameter and is required for this inverse modeling to restore the image quality. 2. No prior information is required for the image enhancement method. This technique uses qualitative subjective criteria to enhance image quality. No physical model is required to produce visually pleasant images from the original images. This kind of scheme is not very complex and simpler in implementation. These are relatively better performances as well.

V.2.5 Underwater detection by ultrasound

Sonar is the preferred choice of sensor for imaging under water due to its long range and robustness to turbidity. Optical cameras are also popular sensors to use, but they are limited in range due to the attenuation of light in water. The different types of sonar have different strengths and weaknesses most importantly with regards to range, field of view and beam resolution.

Forward Looking Sonar (FLS) is, as the name suggests, a sonar mounted to look forward. There can be several variants of FLS. For the active case the principle of operation is that the sonar insonifies a scene with an acoustic wave. The intensity of the acoustic return is then sampled as a function of range and bearing (r,θ). In the 2D case images built up from the acoustic returns are displayed in Cartesian coordinates. For an Autonomous Underwater Vehicle (AUV) the FLS is used to feed back information to control the vehicle. An example of use for FLS on AUV is for monitoring a harbour. While operating underwater the AUV will have to respond to its environment. An example of information the AUV might need is: Are there any obstacles that the AUV needs to avoid? Or, is there a diver in this harbour area? For answering both these questions it is useful for the AUV to be able to recognize what it ”sees” in its acoustic image from the FLS. Both resolution limitations and the noise characteristics of sonar make automatic classification of FLS images very challenging. When we compare acoustic images produced by FLS to high resolution sonar system such as Synthetic Aperture Sonar (SAS) it is clearly more demanding to classify the lower resolution images. An example of what a SAS image looks like can be seen in Figure V.7 (left) where a tripod is depicted. An example of a FLS image of a boat hull is seen in Figure V.7 (right). The SAS image appears more photo-like than the FLS.

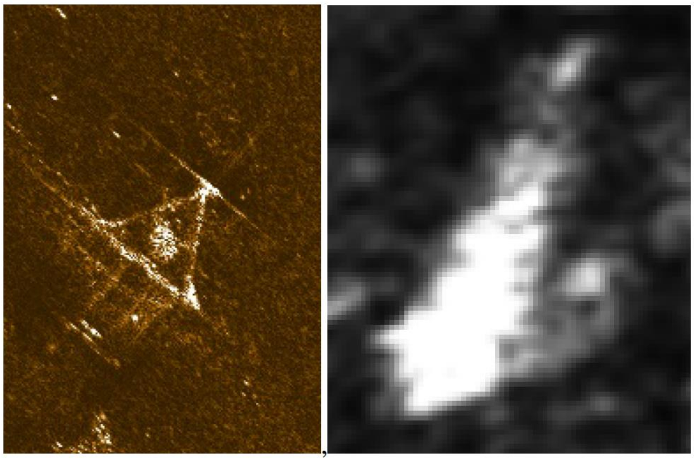


Figure V.7: Left: SAS image depicting a tripod, image from Right: FLS image of a boat hull from.

Being able to describe a scene in words as a ”semantic map” has also proven useful to solve mobile robotics tasks for ground robots. To create semantic information for our map, we need to detect and recognize our objects of interest. This work focuses on the object recognition task in FLS. In this work, we explore the use of Deep Learning methods (DL) developed for computer vision to solve the acoustic image recognition problem in FLS images. Our target application is using FLS on an AUV to monitor an area such as a harbour. Collection and annotation of sonar data is difficult and often the amounts of training data is limited. This rules out the use of more data-hungry DL methods. One approach commonly used when limited amount of training data is available, is a method called transfer-learning. We will use a Convolutional Neural Network (CNN) trained on the ImageNet database to extract generic features. Such features can be used to solve a variety of image recognition tasks as shown in for example. Our goal here is to learn good features from limited amounts of data. This can avoid problems such as redundancy, bias or low discrimination power that can arise with hand-crafted features. The contribution of this work is a new approach for object recognition in FLS images that makes use of features learned from a pretrained CNN. Our hypothesis is that CNNs designed for feature extraction for images can work well even for feature extraction for low resolution ”blob-like” images.

We compare the new classifier with a classifier designed by using hand-crafted features. We show that the DL approach of transfer learning can be a good alternative to using hand-crafted features. We also analyse how the size of the training set affects the test accuracy and find that it should be considered when designing a CNN-based classifier. This paper is organized as follows. Section II provides a background on object recognition in sonar. Section III describes the method used and our system design with the new CNN-based classification.

V.3 Scanning Acoustic Microscope (Defect detection)

V.3.1 Introduction

Ultrasound refers to a sound wave having a frequency (> 20 kHz) higher than the values a human being can hear. Wild animals such as bats and dolphins use acoustic waves to detect and locate objects and obstacles and catch prey in a dark environment. Acoustics are not only utilized in military, scientific, and industrial fields, and medical imaging but also in new and efficient evaluation techniques for quantitatively characterizaing the physical properties such as strain, distance, and viscoelasticity of elastic and hard materials.

Interest in acoustics began very early in the sixth century BC when Pythagoras’ proposed the mathematical properties of the strings on instruments. The application of acoustic microscopy was initiated by Sokolov in 1949 and a breakthrough was achieved in the early 1970s in high-resolution imaging for investigating the internal structure of nontransparent solids. Acoustic microscopy has been substantially improved by Quate at Stanford University in 1985 for achieving an accurate damage or defect detection in microelectronics and early diagnosis of mutations in the tissues or organs as a non-invasive analytical method.

Optical microscopy, X-ray computer-aided tomography, infrared thermography, and scanning acoustic microscopy (SAM) are the commonly utilized techniques owing to their short time consumption and low sample pretreatment requirement as compared to electron. The most striking advantage of SAM is its ability to inspect the sample subsurfaces layer by layer simultaneously with an excellent penetrating power of the ultrasonic waves while scanning the material surface. Thus, three-dimensional (3D) tomography is possible for inner structure integration by using complex and sophisticated algorithms of signal processing with a lateral two-dimensional scan. At present, the ultrasonic technique has become a robust method for non-invasive material evaluation and defect detection for conducting a quality assessment and metrology in a short time and with high efficiency.

V.3.2 Basic principles of SAM

A SAM consists of an ultrasonic emitting transducer, a mechanical scanner and an image processor. The transducer plays an important role as a lens that delivers and focuses the acoustic wave generated by the piezoelectric array and as a detector that accepts the echo reflected from the sample. The ultrasonic pulses (acoustic waves) are generated by a piezoelectric transducer comprising of a magnet and a radiofrequency coil. The transducer crystals are made up of lithium niobate ($LiNbO_3$) single crystals, lead magnesium niobate-lead titanate (PMN-PT) single crystals, quartz or piezoelectric ceramics for frequencies below 100 MHz. Above this limit, a zinc oxide (ZnO) crystal is often used. The piezoelectric transducer comprises a lens, a matching layer, and active piezoelectric array elements that are connected to electrical lines. The acoustic signal oscillates at its own frequency when it receives an electronic intermittent pulse excitation and is delivered via a sapphire cylinder (Al_2O_3) to the lens. The flat wavefront focuses along the z-axis after refraction at the lens/coupling medium interface. Refraction occurs at the interface because of the different velocities of the sound waves within the two materials. Since most materials have a higher speed than a water coupling medium, the focal length is shortened. This phenomenon can be caused by refraction. Acoustic coupling via a water medium between the lens (transducer) and the sample (object) is very important for an efficient delivery of the acoustic waves.

Variations in the acoustic impedance at the interfaces of the tested material lead to the reflection and scattering of the ultrasonic waves. A signal is acquired after the interaction of the acoustic pulse with the sample in the reverse order. The reflected signal (echo) is collected with a diverging form, but is transformed into a flat £wave after a lens and is then converted into an electrical signal with a spatial coordinate. The electrical voltage is amplified and digitized using a image processor.

Using a raster scanning motion of the transducer on the sample surface, the acoustic echo signals are collected at each point and the depth information is recorded in a histogram. 3D images can be reconstructed from all recorded histograms for the region of interest. Various scanning pathways are possible for handling the algorithms of the transducer. More details of the scan motions are explained later. A time duration must be recorded when a series of acoustic excitations occurs owing to the transit time (t) of the reflections or echoes in the sample. The repetition rate is limited by a processor-controlled process depending on the frequency used. The lower the frequency, the longer are the pulses, but the attenuation decreases.

V.3.3 See-through inspection using SAM

Figure V.8 shows the schematic of a reflection mode SAM applied to a material that includes various defects such as grain boundaries, cracks, delamination, voids, bubbles, and contaminants. The SAM uses an ultrasonic pulse from the transducer to probe the sample and monitors the reflected echoes from a series of time delays of the pulses induced after interacting with the internal structure. These delays depend on the acoustic properties of the material, such as its density and attenuation that is determined by the acoustic impedance of the material.

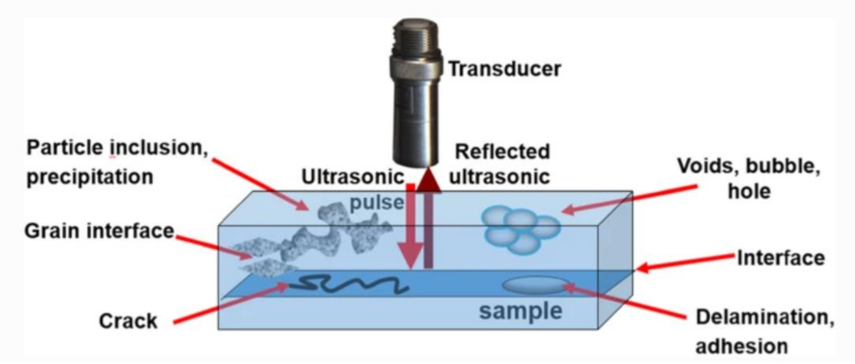


Figure V.8: Probing pulse and the reflected echo detected from the internal structure of a sample consisting of a variety of defects and inclusions

The ultrasonic pulse generated from the transducer is delivered to the sample via a water medium. During this process, some fraction of the pulse bounces off the surface owing to the change in the acoustic impedance (Fig. 1). The ratio of the reflected waves to the transmitted waves is determined from the impedance mismatch. The acoustic impedance (Z) is defined as the amount of resistance an ultrasound beam experiences as it passes through a material and is expressed as follows:

$$Z = \rho \times c$$

Where ρ material density (in kg/m^3), c is the velocity of sound within the material (in m/s) and Z is related to the acoustic hardness of the material. Therefore, if the density of the material increases, its impedance increases. Similarly, when the velocity of sound in the material increases, its impedance also increases. When an ultrasound beam encounters an interface at normal incidence, some of the energy is reflected (reflection) off its surface and some amount is transmitted (transmission) through the interface. The amplitude of the reflected echo depends on the acoustic impedance at the interface or the boundary between the two materials. We can intuitively understand that most of the sound is reflected when the difference in acoustic impedance is large. The ultrasound intensity or the amount of the reflected and transmitted waves can be evaluated using the reflection coefficient, *R*, and the transmission coefficient, *T*, respectively.

$$R = \frac{I_R}{I_0} = \frac{(Z_2 - Z_1)^2}{(Z_2 + Z_1)^2}$$

$$T = \frac{I_T}{I_0} = 1 - R = \frac{4Z_2 Z_1}{(Z_2 + Z_1)^2}$$

Where Z_1 and Z_2 are the acoustic impedances of 1st and 2nd material.

For example, the amount of energy or intensity that is reflected at the water-to-polypropylene interface, using the acoustic impedance values of water and polypropylene of 1.48 and 2.48, respectively, becomes R = 0.0638. Thus, 6.38% intensity is reflected and 93.6% is transmitted.

The higher the acoustic impedance mismatch at the interface, the larger is the intensity of the reflection or brighter is the reflected image. The energy loss is represented as an attenuation while the ultrasound beam progresses through a medium. In practice, we should also consider three major attenuation factors other than reflection and transmission, namely, diffraction, scattering, and absorption of the ultrasound beam.

An acoustic pulse passes through three boundaries, namely, the surface, defects, and the bottom, and the time-of-flight (t) of the pulse reflected from each surface provides the depth information of the sample. A series of reflected waveforms are recorded in real time with a phase histogram, called, the A-scan of the reflection mode SAM. The ultrasonic wave experiences a phase inversion at the interface of the high and low acoustic impedance region, thus resulting in a minimum of three consecutive amplitudes of the signal. The (+) and (−) phase signals should cross the baseline (vertical dotted line in Figure V.8). The SAM instrument automatically detects a histogram with a phase often caused by irregularities inside the sample such as delamination, void, or a hole. The ultrasonic wave propagation in the acoustic microscope results in the wave front snapshots calculated along the time-of-flight of the echo.

Figure V.9 shows the results of qualitative acoustic imaging applied to the detection of adhesion defects constitutes one of the most important applications of this technique.

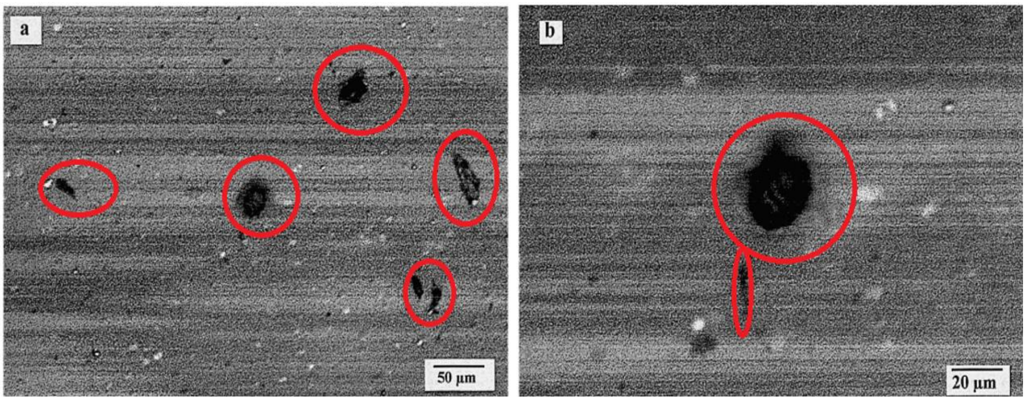


Figure V.9: Acoustic image at 600 MHz of an Al/PET system
(a) 50 x35 µm², (b) 20 x 14 µm).

The flows that manifest themselves by the creation of an air bubble at the interface, are at the origin of a very strong contrast (intense reflection of acoustic waves) caused by the significant variation of the acoustic impedance between two materials that are no longer in direct contact. An example is given in Figure II-4 of the detection of delaminations of the surfaces of a structure Aluminum deposited on polyethylene trephthalate (Al/PET) is one of the defects among which studied and the size and distribution of these interfacial adhesion defects are determined by acoustic imaging.

V.4 Sonochemistry

V.4.1 Introduction

The term "sonochemistry" is used to describe the chemical and physical processes occurring in solution through the energy brought by power ultrasound. The effects of ultrasound are the consequence of the cavitation phenomenon, which is the formation, the growth and the collapse of gaseous micro bubbles in liquid phase (Figure v.10). Ultrasound is propagated through a series of compression and rarefaction waves in the liquid medium. When the acoustic power is sufficiently high, the rarefaction cycle exceeds the attractive forces of the molecules of the liquid and cavitation bubbles of a few micro-meters in diameter are formed. Small amounts of vapor or gas from the medium enter in the bubble during its expansion phase and is not fully expelled during compression phase. The bubbles grow over the period of a few cycles to an equilibrium size for the particular frequency applied. The intense local effects (mechanical, thermal and chemical) due to the sudden collapse of those bubbles are at the origin of all the applica tions of sonochemistry.

In water, at an ultrasonic frequency (*f*) of 20 kHz, each cavitation bubble collapse represents a localized hot-spot, generating temperatures of about 5,000 K and pressures superior to 1,000 bars (Figure V.10). Many factors can affect the cavitation and the results of a sonochemical reaction: the acoustic power, the frequency, the hydrostatic pressure, the nature and the temperature of the solvent, the used gas and even the geometry of the reactor. The potential of sonochemistry is often directly connected to the choice of the sonochemical parameters or experimental conditions.

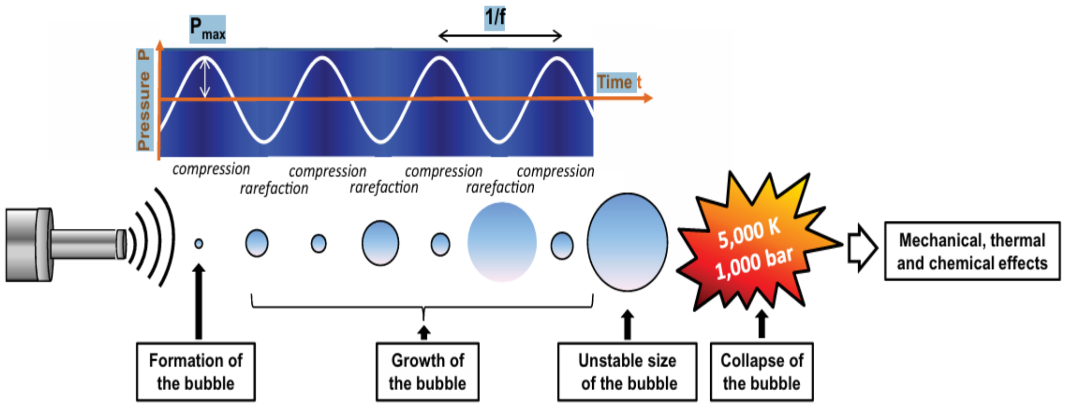


Figure V.10: Schematic representation of the acoustic cavitation phenomenon.

For example, the frequency is an essential parameter (Figure V.11). Indeed, even if the whole mechanism is not elucidated yet, it is usually accepted that, in water, low frequencies (20–80 kHz) lead preferentially to physical effects (shockwaves, micro-jets, micro-convection, etc.). On the contrary, high ultrasonic frequencies (150–2,000 kHz) favor the production of hydroxyl radicals (HO•) through the local hot spots produced by cavitation, mainly leading to chemical effects. In a broad outline, it is possible to identify two great families of power ultrasound applications in chemistry based either on sonophysical effects or sonochemical effects. Conditions obtained in a medium sub mitted to ultrasound are accountable for a large number of physicochemical effects as increase in kinetics of chemicals reactions, changes in reaction mechanisms, emulsification effects, erosion, crystallization, precipitation, etc.

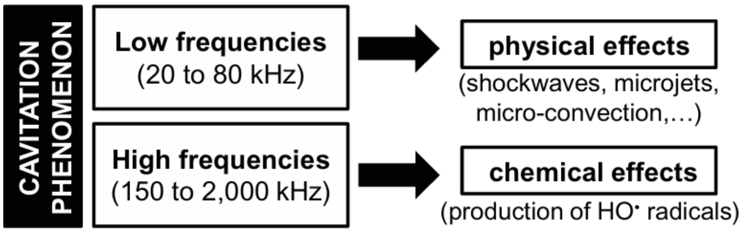


Figure V.11: Predominant effects in water as a function of the frequency range.

The design of organic reactions, material preparations or other chemical processes under ultrasound requires a rigorous methodology and the complete report of all sonochemical parameters and experi mental details. This latter point is essential for two reasons:

- Based on the available literature, the comparison between the obtained results and the wide spread of mechanistic explanations is not always trivial;
- In the absence of these precautions, it could be really difficult (and even impossible in many cases) to reproduce an experiment from the literature.

V.4.2 Chemistry Based on Ultrasonic Waves

By definition, an ultrasonic wave is a sound wave belonging to the range between 20 kHz and 200 MHz that can be subdivided into two distinct regions: power ultrasound and diagnostic ultrasound (Figure V.12). At lower frequencies, greater acoustic energy can be generated to induce cavitation in liquid medium (sonochemistry). Ultrasonic frequencies above 2 MHz do not produce cavitation. This range is particularly used in medical imaging (diagnostic ultrasound). The sound is a wave generated by a mechanical vibration that travels due to the elasticity of the surrounding environment through longitudinal waves (straight line).

- The sound waves do not propagate in vacuum.
- The sound propagation is an elastic deformation (reversible): there is no transport of material.

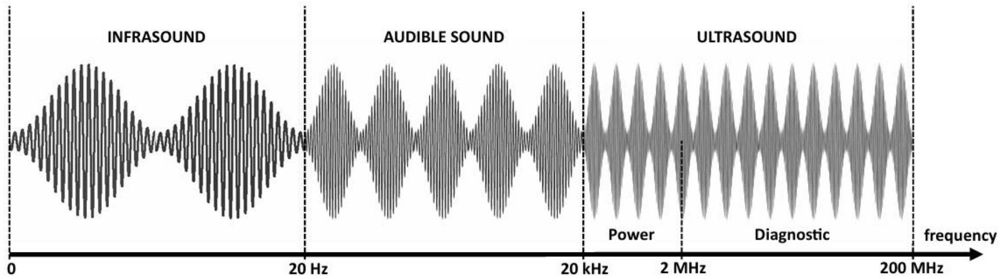


Figure V.12: Frequency ranges of sound

To be described, a sound wave is often considered as a sinusoidal plane wave which is particularly characterized by the following properties:

- The frequency (*f*) in Hertz (Hz) or its inverse, the period (*T*) in seconds (s):

$$f = \frac{1}{T}.$$

- The wavelength (l) usually expressed in nanometers (nm) or its inverse, the wave number in cm^{-1}.

$$\bar{\nu} = \frac{1}{\lambda}.$$

- The alternative pressure wave is also characterized by its amplitude (P). The temporal evolution of the amplitude *P(t)* follows the simplified Eq.

$$P(t) = P_{\max} \times \sin(2\pi t + \Phi),$$

Where P_{max} is the maximal amplitude, *t* is the time and *Φ* is the phase.

Figure V.13 schematically represents the wave with the evolution of the amplitude as a function of the time. We do not give more details on the determination/calculation of the parameters such as amplitude or phase since they should not be very useful for the chemist who wants to develop sonochemical applications.

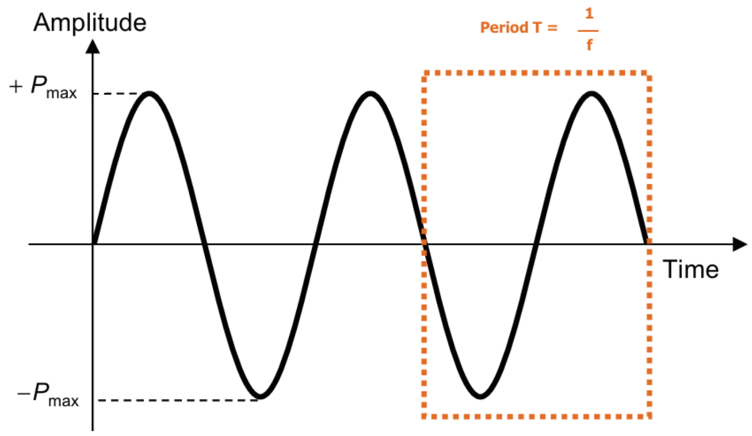


Figure V.13: Representation of the sinusoidal plane wave of sound.

- The sound pressure or acoustic pressure, expressed in Pascal (Pa), is the local pressure deviation from the ambient pressure caused by a sound wave. It could be measured using a microphone in air or a hydrophone in water.
- The speed of sound, denoted c and expressed in meter per second ($m.s^{-1}$), is the distance travelled per unit time by a sound wave propagating through an elastic medium.

V.5 Thermoacoustics

V.5.1 Introduction

Thermoacoustics is a multidisciplinary science dealing with the conversion between the acoustic energy and thermal energy. The expression “thermoacoustic” was first proposed in the 1970s by N. Rott, who derived a linear theory of thermoacoustics. Nevertheless, the first observations of the thermoacoustic effect can be dated back more than 200 years ago, when in 1777 B. Higgins discovered the relationship between the acoustic oscillations excited by the hydrogen flame and the placement of this hydrogen flame inside a large tube. In 1850, Sondhauss experimentally studied the phenomenon of heat-generated sound observed for centuries by the glassblowers, when blowing a hot bulb at the end of a cold narrow tube. Only nine years later, in 1859, Rijke modified the Higgin’s tube and placed a heated wire screen in the lower part of an open-ended tube, which led to strong acoustic oscillations. An important improvement of Sondhauss' tube was done in 1962 by Carter and co-works. They enhanced sound generated from tube by placing inside it a new structure-stack. The idea of Carter et al., further developed by Feldman, led to the construction of the first thermoacoustic engine, which produced 27 W of acoustic power from 600 W of heat. It was one of the most crucial steps in the progress of modern thermoacoustic technology, which began a lot of theoretical and experimental research on the thermoacoustic phenomena and the thermoacoustic devices.

V.5.2 Thermoacoustic devices

Thermoacoustic devices rely on the thermal interactions between the compressible fluid undergoing an acoustic oscillation and the porous structure that is placed in the resonator. As the fluid oscillates in the porous structure, it experiences changes in pressure, displacement, and temperature. On the one hand, the temperature changes come from adiabatic expansion and compression by acoustic wave, on the other, they are a result of the heat transfer between the fluid and the porous structure. There are two main type of porous medium used to exchange heat with the working fluid-stack and regenerator. In the stack, the hydraulic radius (rh)(rh) of a typical pore is greater than the thermal penetration depth (δk)(δk), defined as a distance across which heat can diffuse through the gas in a time that is related to the acoustic period of the acoustic oscillation $(T=2\pi/\omega)$(T=2π/ω). Because in the stack $rh/\delta k \geq 1$rh/δk≥1, the portions of the fluid that are close to the solid are simply compressed and expanded isothermally by acoustic wave, while the portions of the fluid at distances of a few thermal penetration depth are compressed and expanded adiabatically. However, in the stack are also fluid parcels that are about thermal penetration depth from the solid. These parcels have sufficient thermal contact to exchange some heat with the porous material and as well insufficient thermal contact to introduce a time shift between the temperature and the pressure oscillation. This imperfect thermal contact between the stack and the fluid is necessary for the heat pump process. The operation of the thermoacoustic devices with the stack requires pressure and displacement of a fluid parcel in the phase, thus these devices are also called standing-wave engines or refrigerators. Whether the thermoacoustic device works as engine or refrigerator (heat pump) depends on the relative phasing of motion and the heat transfer direction. The engine converts thermal energy into acoustic energy, while the refrigerator utilizes acoustic energy to transport heat from a low-temperature source to a high-temperature source. A more detailed explanation of the operating principles of the thermoacoustic devices is available in Swift's review article.

In the recent years, thermoacoustic devices have been receiving growing interest in research for their many advantages, such as using environmentally friendly working fluids, possibility of utilizing renewable energy sources, having few moving components, simple construction, and high reliability. For example, the main components of a typical thermoacoustic refrigerator are a stack, an acoustic driver, in most cases an electrodynamic loudspeaker and the heat exchangers for high-temperature and low-temperature heat source (Figure V.141). However, wide application of thermoacoustic devices in the industry is still limited due to poor efficiency, which remains one of the problems facing this relative new technology. Therefore, thermoacoustic refrigerators and engines have been extensively studied experimentally as well as theoretically.

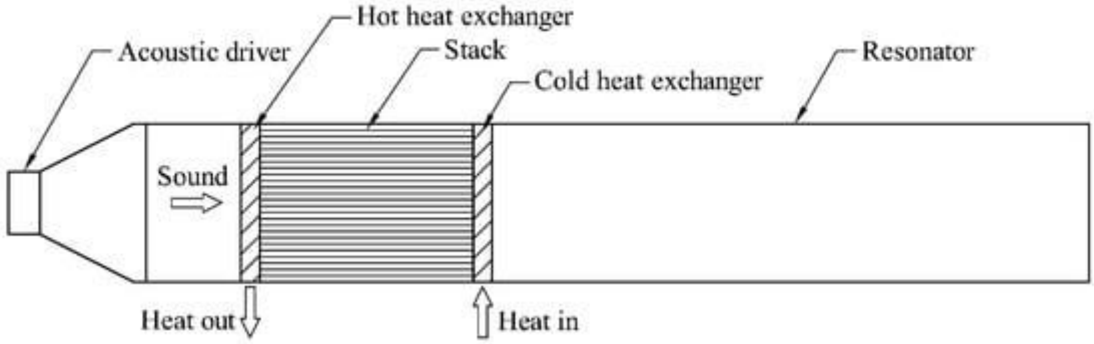


Figure V.14: Schematic illustration of the thermoacoustic refrigerator with standing wave.

Most of the current investigations for thermoacoustic devices with standing wave have been done on the stack, which especially concern stack geometry, position, and length. The stack can be made of spiral plate, parallel plate, circular pores, pin arrays, or triangular pores. The two first types are the most often used, studied and analyzed.

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