Basics of Ultrasound

Principles of Medical Imaging

Prof. Dr. Philippe Cattin

MIAC, University of Basel

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Introduction

Abstract

Why Ultrasound

- Ultrasound (US) is the most widely used imaging technology worldwide
- Popular due to availability, speed, low cost, patient-friendliness (no radiation)
- Applied in obstetrics, cardiology, inner medicine, urology,...
- Ongoing research to improve image quality, speed and new application areas such as intra-operative navigation, tumour therapy,...
Applications in Obstetrics  (5)

- Follow fetal development
- Detect pathologies

Fig. 4.1: Two-dimensional B-mode Ultrasound image of a fetus

Fig. 4.2: Three-dimensional image of the same fetus ~5 months after conception

Applications in Cardiology  (6)

- Blood flow in vessels (Doppler US)
- Contraction, Rhythm
- Blood flow in the heart (defects on wall muscle, valve defects)
- Assessment of cardiac perfusion
Applications in Inner Medicine

- Gallstone
- Perfusion of renal transplant

Fig. 4.5: Gallstone (red arrow) within the gallbladder produces a bright surface echo and causes a dark acoustic shadow (S)

Fig. 4.6: Perfusion Doppler image of a renal transplant
Applications in Musculoskeletal System

- Visualisation of tendons, ligaments
- Investigation under movement is possible → simplifies the detection of ruptures, obstructions, ...

![Fig. 4.7: The arrows show the large gap of the rupture Achilles tendon](image)

![Fig. 4.8: US image of ISS astronaut Mike Fincke's biceps tendon, where "D" denotes the deltoid muscle and "T" is the proximal intracapsular end of the long biceps tendon](image)

Applications of Ultrasound

Elastography

- US Elastography is often used to classify tumours. Malignant tumours are 10 to 100 times stiffer than the normal soft tissue around.

![Fig. 4.9: Elastogram (of a breast) indication a mass with a high probability of being malignant tumour](image)
**Advantages and Disadvantages of Ultrasound**

### Advantages
- *No ionising radiation*
- *Real-time* imaging (functional examinations)
- *Small*, transportable equipment
- Comparatively low cost

### Disadvantages
- Moderate to *poor image quality*
- Image interpretation is *difficult* and operator dependent
- Acoustic window
- *Limited* penetration depth (attenuation, total reflection)
- *Shadows* by dense objects

---

**Challenges for Image Analysis**

- Artefacts: speckle pattern, attenuation, shadows, contours parallel to the beam direction
- Lateral resolution depends on the depth

---

*Fig. 4.10: SmartPhone with Ultrasound capability*
*Fig. 4.11: Application scenario*
*Fig. 4.12: US image of the heart chambers*
*Fig. 4.13: US image of a gallstone*
Challenges for Image Analysis (2)

Sound propagates at different speed in different tissues

- Estimation of distances difficult (1540 m/s assumed in Ultrasound devices)
- Distortion of contours in Ultrasound images

<table>
<thead>
<tr>
<th>Material</th>
<th>c [m/s]</th>
</tr>
</thead>
<tbody>
<tr>
<td>Water</td>
<td>1480</td>
</tr>
<tr>
<td>Blood</td>
<td>1560</td>
</tr>
<tr>
<td>Brain</td>
<td>1530</td>
</tr>
<tr>
<td>Fat</td>
<td>1476</td>
</tr>
<tr>
<td>Muscle</td>
<td>1568</td>
</tr>
<tr>
<td>Bone</td>
<td>2800 – 4100</td>
</tr>
<tr>
<td>Air</td>
<td>330</td>
</tr>
</tbody>
</table>

Tab. 4.1: Different important sound speed in tissue

Fig. 4.14: Different sound speeds distort distance measurements

Challenges for Image Analysis (3)

Segmentation in the presence of noise (speckle patterns are in a strict sense not noise but interferences of reflections) and artefacts.

Fig. 4.15: Segmentation example 1
Fig. 4.16: Segmentation example 2
Myocardial Contrast Echocardiography

In Myocardial Contrast Echocardiography (MCE) a contrast agent micro bubbles is injected into the circulation. These bubbles produce a high-contrast response visible as bright sport in the movie.

Procedure

1. Contrast agent is injected
2. Heart is imaged with the Ultrasound system
3. By increasing the Ultrasound energy the bubbles in the Ultrasound plane are destroyed (blue frame)
The time it takes to refill the myocardium with bubbles is a measure of muscle perfusion (health of the coronary arteries).

**Problem**

Analysing and evaluating these movies by hand is a time consuming and error prone task.

**Myocardial Contrast Enhancement (2)**

The developed software allows to analyse a US movie sequence in a matter of minutes rather than one to two hours.
History

1877: Lord Raleigh - "Theory of Sound"
1880: Pierre & Jacques Curie - Piezoelectric effect
1914: Langevin - First Ultrasound generator using piezoelectric effect
1928: Solokov - Ultrasound for material testing
1942: Dussik - First application of Ultrasound in medical diagnostics
... different medical applications (gall stones, tumours)
End of 1960’s: Boom of Ultrasound in medical diagnostics

Fig 4.18: The bat use Ultrasound for navigation
Fig 4.19: Edlet and Hertz's echocardiographic trace of the mitral valve (1950)

History (2)

1957: Pan-Scanner - The transducer rotated in a semicircular arc around the patient
1957: Scan converter allowed for the first time to use the upcoming computer technology to improve US
1957: A state-of-the-art Ultrasound console

Fig 4.20: Pan-Scanner - The transducer rotated in a semicircular arc around the patient
Fig 4.21: Scan converter allowed for the first time to use the upcoming computer technology to improve US
Fig 4.22: A state-of-the-art Ultrasound console
Properties of Ultrasound

The frequencies of medical Ultrasound waves are several magnitudes higher than the upper limit of human hearing [http://en.wikipedia.org/wiki/Human_hearing].

Fig. 4.24: Approximate frequency ranges of sound
Common Sound Frequencies

<table>
<thead>
<tr>
<th>Sound</th>
<th>Frequency</th>
</tr>
</thead>
<tbody>
<tr>
<td>Adult audible range</td>
<td>15 – 20,000 Hz</td>
</tr>
<tr>
<td>Range for children's hearing</td>
<td>up to 40,000 Hz</td>
</tr>
<tr>
<td>Male speaking voice</td>
<td>100 – 1,500 Hz</td>
</tr>
<tr>
<td>Female speaking voice</td>
<td>150 – 2,500 Hz</td>
</tr>
<tr>
<td>Standard pitch (Concert A)</td>
<td>440 Hz</td>
</tr>
<tr>
<td>Bat</td>
<td>50,000 – 200,000 Hz</td>
</tr>
<tr>
<td>Medical Ultrasound</td>
<td>2.5 – 40 MHz</td>
</tr>
<tr>
<td>Maximum sound frequency</td>
<td>600 MHz</td>
</tr>
</tbody>
</table>

Tab. 4.2: Common sound frequencies and frequency ranges

Physical Principles of Ultrasound

Ultrasound imaging differs in three fundamental aspects from other imaging methods:

1. Ultrasound is a non-ionising longitudinal wave
2. The reflected signal is recorded rather than the transmitted part (unlike e.g. X-ray)
3. Ultrasound images tissue boundaries instead of e.g. density information, see Fig. 4.25

Comparing Ultrasound with other imaging modalities, such as CT or MRI, thus requires image processing.
**Acquisition Principle**

Ultrasound measures the time a pulse takes to travel from the transducer to the reflecting surface and back, see Fig 4.26 → time-of-flight.

- The depth can be calculated, if the sound velocity $c$ of the tissue is known.
- The magnitude of the echo is coded as gray value on the display.

![Diagram of ultrasound waves](image)

**Propagation**

Ultrasound is a cyclic longitudinal pressure wave and requires a medium (gas, liquid, solid).

Ultrasound waves behave according to the conventional laws associated for light waves except that they require a medium.
Propagation (2) (27)

1. The soundwave applies a force from the left to particle '1'
2. Particle '1' acquires a speed of \( v \) about its center and moves to the right by compressing and stretching its connecting bonds
3. Particle '2' now receives that force from '1' and is displaced to the right
4. Now that particle '1' has transferred its force it moves back to its equilibrium position
5. The displacement of particle '2' is slightly smaller than '1' due to energy loss (heat) within the molecular bond causing the soundwave to slowly fade out

This vibration process of compression and rarefaction is repeated for the remaining particles.

Fig. 4.29: Propagation of sound energy as elongation about a center of equilibrium (particle velocity about its center is \( v \))
Propagation (3) (28)

The propagation of a soundwave can be visualised, Fig 4.30, as a sine wave with a
- Wavelength $\lambda$,
- Frequency $f$, and
- and an Amplitude $A$

that characterises the transmitted Ultrasound wave in the tissue.

Ultrasound Characteristics (29)

Sound particle velocity $v$
- Sound particle velocity $v$ [mm s$^{-1}$] is the velocity of the material particles as they oscillate about their equilibrium.

Acoustic Pressure $p$
- The Acoustic Pressure $p$ is caused by the pressure changes and measured in Pascal [Pa]

Frequency $f$ and Wavelength $\lambda$
- The Frequency [s$^{-1}$] and the Wavelength [mm] are related with $\lambda = c/f$, where $c$ is the speed of sound (propagation velocity) in the respective tissue.

Propagation Velocity $c$
- The Propagation Velocity $c$ is the speed with which the soundwave travels through the medium.
- This velocity is tissue specific and often only approximately known. The stiffer the springs and the smaller the particle masses, the higher to propagation velocity.
- Propagation velocity, wavelength and frequency are related by
  $\lambda = \frac{c}{f} \ (4.1)$
- The propagation velocity increases from gases to liquids
and is highest in solids.

**Ultrasound Wavelength Change**

As we have seen, the Wavelength $\lambda$, Frequency $f$ and Propagation Velocity $c$ are related by

$$c = \lambda f \quad (4.2)$$

When the Ultrasound wave travels from one medium to the other, the velocity changes.

This change of velocity thus induces a change of the Wavelength and not the Frequency of the Ultrasound wave, see Fig 4.31.

Fig. 4.31: If the speed of sound changes in a medium, the wavelength changes as well.
Ultrasound Characteristics (31)  (2)

Amplitude $A$

- The maximum Amplitude coincides with the compression peak, see Fig 4.30. Reducing the power level results in a smaller amplitude.

Power

- The Power is the rate of sound energy transfer into the tissue and is measured in watts [W = Js$^{-1}$]

Intensity

- The Intensity is a measure of power per unit area and is commonly measured as $Wcm^{-2}$.

Modulus of Elasticity (32)

The rate of transfer from one molecule to the other depends on the molecule’s reluctance to motion and the density. The wave velocity (speed of sound) $c$ are connected through Young’s Modulus of Elasticity $E$.

$$c = \sqrt{\frac{E}{\rho}} \quad (4.3)$$

where $c$ is the speed of sound, $\rho$ the material density and $E$ Young’s Modulus.

- The higher the modulus $E$ (stiffer springs) and lower molecule masses, the higher the velocity
- The modulus of elasticity is inversely proportional to the compressibility $1/E$
- The modulus $E$ is measured in pressure units e.g. [GPa]
  - Fat $E_{fat} = 2.0 \text{ GPa}$
  - Soft tissue $E_{st} = 2.5 \text{ GPa}$
  - Bone $E_{bone} = 25 \text{ GPa}$
### Acoustic Impedance \( (33) \)

When pressure \( p \) is applied to a molecule it will move exerting a pressure on an adjacent molecule. Acoustic pressure increases with particle velocity but is also depends on properties of the medium. The equation is similar to electrical resistance:

\[
\begin{align*}
\text{Acoustic Impedance} \quad Z & \quad \text{Electronic Resistance} \quad R \\
Z &= \frac{p}{v} \quad (4.4) & R &= \frac{U}{I} \quad (4.5) \\
v &= \frac{p}{Z} \quad (4.6) & I &= \frac{U}{R} \quad (4.7) \\
p &= v \times Z \quad (4.8) & U &= I \times R \quad (4.9)
\end{align*}
\]

The acoustic impedance \( Z \) is measured in \([\text{kg m}^{-2} \text{s}^{-1}]\) often shortened to \(\text{rayl}\).

### Acoustic Impedance (2) \( (34) \)

The Acoustic Impedance \( Z \) is also related to the Modulus of Elasticity \( E \).

- The stiffer the bonding, the greater the pressure exerted by a molecule moving at a particular velocity, so

\[
p = Zv \quad (4.10)
\]

A material having a great springiness (low \( E \) value) has high molecular motion and will absorb sound energy and less will be transferred to the next molecule, so

\[
Z = \frac{E}{c} \quad (4.11)
\]

combining Eq. 4.10, 4.11 and 4.3 yields

\[
Z = \rho c = \sqrt{E\rho} \quad (4.12)
\]
Properties of Different Materials

<table>
<thead>
<tr>
<th>Material</th>
<th>Velocity $c$ [m/s$^{-1}$]</th>
<th>Density $\rho$ [kg m$^{-3}$]</th>
<th>Acoustic Impedance $Z$ [Rayl = kg m$^{-2}$s$^{-1}$, $\times 10^6$]</th>
</tr>
</thead>
<tbody>
<tr>
<td>Air</td>
<td>330</td>
<td>1.3</td>
<td>0.00043</td>
</tr>
<tr>
<td>Fat</td>
<td>1470</td>
<td>970</td>
<td>1.42</td>
</tr>
<tr>
<td>Castor oil</td>
<td>1500</td>
<td>933</td>
<td>1.40</td>
</tr>
<tr>
<td>Water</td>
<td>1492</td>
<td>1000</td>
<td>1.48</td>
</tr>
<tr>
<td>Soft tissue</td>
<td>1500</td>
<td>&lt;1000</td>
<td>~1.45</td>
</tr>
<tr>
<td>Brain</td>
<td>1530</td>
<td>1020</td>
<td>1.56</td>
</tr>
<tr>
<td>Blood</td>
<td>1570</td>
<td>1020</td>
<td>1.60</td>
</tr>
<tr>
<td>Kidney</td>
<td>1561</td>
<td>1030</td>
<td>1.61</td>
</tr>
<tr>
<td>Liver</td>
<td>1549</td>
<td>1060</td>
<td>1.64</td>
</tr>
<tr>
<td>Muscle</td>
<td>1568</td>
<td>1040</td>
<td>1.63</td>
</tr>
<tr>
<td>Eye lens</td>
<td>1620</td>
<td>1130</td>
<td>1.83</td>
</tr>
<tr>
<td>Bone (compact)</td>
<td>4080</td>
<td>1700</td>
<td>6.12</td>
</tr>
<tr>
<td>Bone (porous)</td>
<td>1700</td>
<td>970</td>
<td>2.5</td>
</tr>
<tr>
<td>Bone marrow</td>
<td></td>
<td></td>
<td>1.65</td>
</tr>
</tbody>
</table>

The different speeds of sound in the tissue makes it difficult to measure accurate depths and distances in the image.

Power and Intensity

\[ P = pv \] (4.13)  \[ P = UI = I^2R \] (4.14)  \[ J/s = W \]

\[ I = P/A \] (4.15)  \[ W/m^2 \]

\[ E = Ic \] (4.16)  \[ J = Ws \]

where $A$ is the area.

- Increasing the energy drives the compression bands closer together → more energy in the tissue and higher oscillation speed
- During travel, energy is lost as heat decreasing the wave amplitude → frequency and wavelength remain for the same tissue
Symbols often used to Characterise Ultrasound

<table>
<thead>
<tr>
<th>Measurement</th>
<th>Symbol</th>
<th>Unit</th>
<th>Clinical Range</th>
</tr>
</thead>
<tbody>
<tr>
<td>Velocity (speed of sound)</td>
<td>$c$</td>
<td>[m/s]</td>
<td>1500 m/s (for soft tissue)</td>
</tr>
<tr>
<td>Wavelength</td>
<td>$\lambda$</td>
<td>[mm]</td>
<td>0.6 – 0.15 mm (for soft tissue)</td>
</tr>
<tr>
<td>Frequency</td>
<td>$f$</td>
<td>[Hz]</td>
<td>2.5 – 10 MHz</td>
</tr>
<tr>
<td>Elastic modulus</td>
<td>$E$</td>
<td>[Pa]</td>
<td>25 GPa (in bone)</td>
</tr>
<tr>
<td>Acoustic impedance</td>
<td>$Z$</td>
<td>[kg/(m²s) = rayl]</td>
<td>1.63 x 10⁶ rayl</td>
</tr>
<tr>
<td>Density</td>
<td>$\rho$</td>
<td>[kg/m³]</td>
<td>1000 kg/m³ (for water)</td>
</tr>
<tr>
<td>Pressure</td>
<td>$p$</td>
<td>[Pa]</td>
<td>0.06 MPa</td>
</tr>
<tr>
<td>Elongation velocity</td>
<td>$v$</td>
<td>[m/s]</td>
<td>&lt; 3.5 cm/s</td>
</tr>
</tbody>
</table>

Interaction with Matter

When Ultrasound interacts with matter, they exhibit the same phenomena as visible light:

- Reflection (specular and non-specular)
- Refraction
- Diffraction
- Attenuation or absorption
Reflection and Refraction  (40)

The laws of wave optics also apply for Ultrasound waves. From energy preservation we know that (neglecting absorption)

\[ I_i = I_r + I_t (4.17) \]

the law of reflection states

\[ \theta_i = -\theta_r (4.18) \]


\[ \frac{\sin \theta_i}{\sin \theta_t} = \frac{\lambda_1}{\lambda_2} = \frac{c_1}{c_2} (4.19) \]

gives the relationship between the sound speeds and angles.

Reflection (2)  (41)

Given the Impedances \( Z_1, Z_2 \) of the two tissues and the angles \( \theta_i, \theta_t \) and the incidence intensity \( I_i \), the reflected and transmitted intensities can be calculated with

\[ I_r = I_i \left( \frac{Z_1 \cos \theta_i - Z_2 \cos \theta_t}{Z_1 \cos \theta_i + Z_2 \cos \theta_t} \right)^2 (4.20) \]

and

\[ I_t = I_i \left( \frac{4Z_1 Z_2 \cos \theta_i \cos \theta_t}{(Z_1 \cos \theta_i + Z_2 \cos \theta_t)^2} \right) (4.21) \]

from Eq \( 4.17 \) we know that

\[ I_t = I_i - I_r (4.22) \]

also holds.
Reflection (3)  (42)

For a sound wave perpendicular to a smooth surface ($\theta_i = 0$), Eq 4.20 can be simplified and yields for the reflected intensity $I_r$

$$I_r = I_i \left( \frac{Z_1 - Z_2}{Z_1 + Z_2} \right)^2$$ (4.23)

and similarly the transmitted wave intensity is given by

$$I_t = I_i \frac{4Z_1Z_2}{(Z_1 + Z_2)^2}$$ (4.24)

- The amount of reflection depends on impedance differences $Z_1, Z_2$.
- Reflection is greatest when the impedance difference is large.

Reflection Example: Air - Fat  (43)

Given:

A sound wave hits an air-fat interface perpendicular to the surface, thus $\theta_i = 0$. The impedance of air is $Z_1 = 0.0004 \times 10^6$ and of fat $Z_2 = 1.42 \times 10^6$.

Result:

The amount of reflected sound intensity can then be calculated using Eq 4.23

$$\frac{I_r}{I_i} = \left( \frac{Z_1 - Z_2}{Z_1 + Z_2} \right)^2$$

$$= \left( \frac{0.0004 \times 10^6 - 1.42 \times 10^6}{0.0004 \times 10^6 + 1.42 \times 10^6} \right)^2$$

$$= 0.999$$

yielding an Ultrasound reflection ratio of 99.9%.

Conclusion:

- Almost all energy is reflected at a tissue-air boundary → Ultrasound is not usable for air-filled cavities such as the lungs,...
Reflection Example: Liver - (44) Kidney

Given:

A sound wave hits a liver-kidney interface perpendicular to the surface, thus $\theta_i = 0$. The impedance of liver is $Z_1 = 1.64 \times 10^6$ and of kidney $Z_2 = 1.61 \times 10^6$.

Result:

The amount of reflected sound intensity can be calculated as in the previous example and yields $I_r/I_i = 0.0085\%$. The rate of transmitted intensity (Eq 4.24) is then given by

$$\frac{I_t}{I_i} = \frac{4Z_1Z_2}{(Z_1 + Z_2)^2} = 0.999915$$

which yields $I_t/I_i = 99.9915\%$

Conclusion:

- Only very little energy is reflected when the impedances closely match → organ boundary not well visible in Ultrasound
**Diffraction**

Diffraction is the bending of Ultrasound into the shadow of a strong absorber. It takes place at the edges of a strong absorber. Although the ultrasonic intensity in the shadow is less than in the incident field, it is not reduced to zero, due to diffraction around the edges of the dense material → image artefacts.

![Fig. 4.35: Diffraction occurs behind the gallstone](image)

**Attenuation or Absorption**

The *intensity* of a sound wave decreases along its direction of propagation. There are two physical processes involved:

- Absorption: responsible for 80% of energy loss
- Beam divergence
Attenuation or Absorption (47) (2)

The decrease in the ultrasonic intensity $I$ with increasing depth $x$ can be characterised by an exponential law

$$I_x = I_0 e^{-\beta x} \quad (4.27)$$

where $I_0$ is the initial sound intensity at $x = 0$ and $\beta$ is a tissue dependent Absorption coefficient, see Fig 4.36.

The rate of energy loss $\beta$ depends on the type tissue and the Ultrasound frequency. Maximum loss is at the relaxation frequency $\omega$.

Attenuation or Absorption (48) (3)

Instead of the absorption coefficient $\beta$ a measure of power loss $\alpha$ is often given:

$$\alpha = 10 \log_{10} \frac{I_0}{I_x} \quad (4.28)$$

where $\alpha$ is the attenuation in [dB] per unit length.

<table>
<thead>
<tr>
<th>Material</th>
<th>Absorption $\alpha$ [dB/(cm MHz)]</th>
</tr>
</thead>
<tbody>
<tr>
<td>Fat</td>
<td>0.6</td>
</tr>
<tr>
<td>Blood</td>
<td>0.18</td>
</tr>
<tr>
<td>Brain</td>
<td>0.85</td>
</tr>
<tr>
<td>Soft tissue</td>
<td>1.0</td>
</tr>
<tr>
<td>Liver</td>
<td>0.9</td>
</tr>
<tr>
<td>Muscle (along fibers)</td>
<td>1.2</td>
</tr>
<tr>
<td>Muscle (across fibers)</td>
<td>3.3</td>
</tr>
<tr>
<td>Eye lens</td>
<td>2.0</td>
</tr>
<tr>
<td>Bone</td>
<td>20.0</td>
</tr>
</tbody>
</table>

Tab. 4.3: Sound absorption of various biological tissues
**Decibel Scale**

Sound power or intensity variations $I_0, I_1$ are compared using the Decibel Scale

$$\text{dB} = 10 \log_{10} \frac{I_1}{I_0} \quad (4.29)$$

Example:

The power loss of an incident source with $I_0 = 1 \text{ W/cm}^2$ and an echo power of $I_1 = 0.1 \text{ mW/cm}^2$ would be

$$\text{dB} = 10 \log_{10} \frac{0.0001}{1} \quad (4.30)$$

$$= 10 \times -4.0$$

$$= -40\text{dB}$$

The half power distance or a reduction of 50% is the distance over which the power is reduced by 3 dB derived from $10 \log_{10} 0.5 = -3.010$.

---

**Attenuation or Absorption**

Higher Ultrasound frequencies are attenuated more than lower frequencies

As a consequence, the required depth (absorption) determines the order of magnitude of the frequencies used in Ultrasound diagnostics, see Fig 4.37.

- **High frequencies** (short wavelengths) → *high spatial resolution* but only *limited penetration depth*
- **Low frequencies** (long wavelengths) → *lower spatial resolution* but *wider depth range*.

Frequencies between 7 – 15 MHz are thus used for superficial structures and frequencies around 2 MHz are used for deeper-lying details e.g. $\approx 25$ cm
Safety of Diagnostic Ultrasound

"Diagnostic ultrasound has proven to be a valuable tool in medical practice. An excellent safety record exists in that, after decades of clinical use, there is no known instance of human injury as a result of exposure to diagnostic ultrasound. Evidence exists, however, to indicate that at least a hypothetical risk for clinical diagnostic ultrasound must be presumed."


The above report explicitly talks of diagnostic Ultrasound. Ultrasound such as HIFU (High-Frequency Focused Ultrasound) does "harm" the patient by destroying tissue. This is of course intended for cancerous tissue or kidney stones.

Bioeffects

Two mechanisms are known to alter biological systems with Ultrasound, the

- thermal mechanism, and
- mechanical mechanism associated with cavitation.
**Thermal Bioeffects**

A major part of the Ultrasound energy emitted during examination is absorbed by the tissue. This absorption results in the generation of heat. If the absorbed energy exceeds the body's ability to dissipate heat, the local temperature will rise. If the temperature is too high, tissue is destroyed.

Measuring the absorption of Ultrasound energy is difficult. It can, however, be estimated solving the differential equation aka bio-heat equation:

\[
cp \frac{\partial T}{\partial t} = k \nabla^2 T + \omega \rho_c \rho_b (T_{\text{blood}} - T) + q_{\text{metabolism}} + \text{SAR} (4.31)
\]

where

- \( \rho, \rho_b \) Density of tissue and blood [kg/m³]
- \( c, \rho_b \) Specific heat of tissue and blood [J/(kg K)]
- \( T, T_{\text{blood}} \) Temperature of tissue and capillary blood [K]
- \( \omega \) Volumetric flow rate of blood per unit mass of tissue [m³/s kg]
- \( q_{\text{metabolism}} \) Rate of metabolic heat generation [W/m³]
- \( \text{SAR} \) Local specific absorption rate [W/m³]

**Non-thermal Bioeffects**

It is known since the 1930's that biological cells can be damaged in the absence of significant heating. This mechanism is linked to the phenomena of acoustic cavitation. Cavitation formation is connected to negative pressure peaks. In the 2 – 20 MHz range a minimum pressure of 15 MPa is required to cause cavitation.

Fig. 4.38: Caviting propeller model in a water tunnel experiment


Fig. 4.39: Cavitation bubble imploding close to surface (e.g. cell) generating a jet (4) of the surrounding liquid

Output Display Standard (56)
Using Thermal and Mechanical Indices

The Ultrasound devices provide a standardised display of real-time indices relating to the potential for Ultrasound-induced bioeffects. Two types of indices are displayed:

- the thermal index (TI) to estimate the temperature increases and
- the mechanical index (MI) estimating non-thermal bioeffects such as cavitation.

The thermal index (TI) is further subdivided into a (TIs) for soft-tissue, (TIb) for bone, and (TIc) for cranial bone. A value of 1 means a temperature increase of 1 K. These estimates are maximum numbers and the real temperature increase is typically significantly lower.