Medical Ultrasound Imaging

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Medical Ultrasound

1. Introduction
2. Main principles
3. Imaging modes
4. Ultrasound instruments
5. Probe types and image formats
6. Bioeffects and safety
7. Emerging technology
Medical Ultrasound uses

- Soft tissues
  - Fetal, liver, kidneys
- Dynamics of blood flow
  - Heart, circulatory system
- Avoid bone, and air (lungs)
Liver: A scan which was considered to be of high quality in the *early 1970s* and which would have been interpreted as supporting the diagnosis of multiple metastases. Wells, Ultrasound imaging, 2006

Metastasis = the spread of a disease from one organ or part to another non-adjacent organ or part. Concerned mainly with malignant tumor cells and infections (Wikipedia)
Liver: A scan made with a modern system, in which a metastasis can just be perceived towards the right side of the patient (i.e., towards the left of the image)
Wells, Ultrasound imaging, 2006
Liver, metastasis: a scan of the same patient, in which this lesion is clearly apparent following the administration of an ultrasonic contrast agent. Wells, Ultrasound imaging, 2006
Medical Ultrasound

- Continuously improving image quality => increased clinical usage
- Relatively inexpensive compared to CT and MR
  - 25% of all imaging exams in hospitals is US
- Many clinical applications
- Requires training and skill
Norway: 30-40 years of development

- 70’s: Started at NTNU, continued R&D since then
  - 2006-2014: Medical Imaging, Centre for Research-based Innovation http://www.ntnu.no/milab

- 80’s: Vingmed Sound

- 1998: GE Vingmed Ultrasound,
  - Now center of excellence for cardiology ultrasound in GE Healthcare
  - More than 2 billion NOK turnover controlled from Horten, Norway

- Other companies:
  - Medistim (quality control in surgery) on Oslo Stock Exchange
  - Sonowand (image-guided surgery based on fusion of ultrasound and MR)
  - Neorad (ultrasound-guided injection of CT contrast agent in vein)
  - Auratech (High quality ultrasound engines/electronics)
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Longitudinal/Compression Waves

Uniform distribution of molecules in a medium

Movement of the piston to the right produces a zone of compression.

Withdrawal of the piston to the left produces a zone of rarefaction.

Alternate movement of the piston establishes a longitudinal wave

Sook Kien Ng, Ultrasound Imaging
Ultrasound: Interaction with tissue

Major interaction mechanisms:

1. Scattering (Imaging + Doppler)
2. Reflection (Imaging)
3. Refraction
4. Attenuation

Lawrence, Physics and instrumentation of ultrasound, 2007.
1. Scattering

A) Rough interfaces

B) Inhomogenous medium (causing speckle in images)

C) Particles (red blood cells)

Burns, Introduction to the physical principles of ultrasound imaging and doppler, 2005
2. Reflection, 3. Refraction

- Due to changes in acoustic impedance $Z = \rho \cdot c$
- Specular
  - Like light striking a glass plate – Snell's law
- Echoes will only be received if beam is near perpendicular
  - Wall of an organ: heart, bladder
- Refraction in some cases – rare anomalies

Burns, Introduction to the physical principles of ultrasound imaging and doppler, 2005
Reflection coefficient

• At any interface between two materials:
• Reflection coefficient:
  – \( r = \) amplitude of reflected/amplitude of incident wave
  – \(-1 \leq r \leq 1\)

  – \( \rho \): density,
  – \( c \): speed of sound, \( c = (E/\rho)^{0.5} \) where \( E \) is elastic modulus
  – \( Z = \rho c = (E\rho)^{0.5} \): acoustic impedance, Rayleigh (Rayl)

\[
 r = \frac{Z_1 - Z_2}{Z_1 + Z_2} = \frac{\rho_1 c_1 - \rho_2 c_2}{\rho_1 c_1 + \rho_2 c_2}
\]
### Reflection coefficient

<table>
<thead>
<tr>
<th>Medium</th>
<th>Density (kg/m³)</th>
<th>Speed of sound (m/s)</th>
<th>Impedance (Rayl)</th>
<th>r</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Air -&gt;</strong></td>
<td>1.3</td>
<td>340</td>
<td>442</td>
<td></td>
</tr>
<tr>
<td>steel</td>
<td>7800</td>
<td>5800</td>
<td>45.24</td>
<td>-0.99998 (180° phase shift)</td>
</tr>
<tr>
<td>water</td>
<td>1000</td>
<td>1490</td>
<td>1.49</td>
<td>-0.981</td>
</tr>
<tr>
<td><strong>Water -&gt;</strong></td>
<td></td>
<td></td>
<td>1.49</td>
<td></td>
</tr>
<tr>
<td>steel</td>
<td></td>
<td></td>
<td>45.24</td>
<td>-0.936</td>
</tr>
<tr>
<td><strong>Muscle tissue -&gt;</strong></td>
<td>1100</td>
<td>1540</td>
<td>1.7</td>
<td></td>
</tr>
<tr>
<td>water</td>
<td></td>
<td></td>
<td>1.49</td>
<td>0.07</td>
</tr>
<tr>
<td>lungs = air</td>
<td>1000</td>
<td>442</td>
<td>7.4</td>
<td>0.9995</td>
</tr>
<tr>
<td>bone</td>
<td>4080</td>
<td></td>
<td>7.4</td>
<td>-0.38 ...</td>
</tr>
</tbody>
</table>

This is why ultrasound imaging works
4. Attenuation

- In water: usually increases with $f^2$
- In tissue: usually increases with $f$

- Typically $\alpha = 0.5$-1 dB/cm/MHz
- Attenuation = $2d \cdot \alpha \cdot f$

- Examples 80 dB attenuation ($\alpha = 0.5$ dB/cm/MHz):
  - $f = 1$ MHz: $d=80$ cm depth, $\lambda = 1.5$ mm => 533 $\lambda$ depth
  - $f = 3$ MHz: $d=26.7$ cm depth, $\lambda = 0.5$ mm => 533 $\lambda$ depth
  - $f = 10$ MHz: $d=8$ cm depth, $\lambda = 0.15$ mm => 533 $\lambda$ depth

- Pavlin, Foster, Ultrasound Microscopy 1998
4. Attenuation

Center frequency shifts with depth

Can show that for a Gaussian spectrum, the center frequency will fall linearly with depth for $f^1$ attenuation (Kuc, 1984, IEEE ASSP)

Simulated in Ultrasim for depths 5, 10, 15, 20 cm
4. Attenuation: Max depth of an image

- Depth where attenuation reaches ~60 dB
  - Attenuation = 2d·α·f => d = Attenuation/(2·α·f)

- Examples: 60 dB attenuation (α = 0.5 dB/cm/MHz):
  - f = 2.6 MHz: d = 23.1 cm depth
  - f = 3 MHz: d = 20 cm depth
  - f = 5 MHz: d = 12 cm depth
  - f = 10 MHz: d = 6 cm depth
4. Attenuation:
PRF - Pulse Repetition Frequency

- \( d = \text{Attenuation}/(2 \cdot \alpha \cdot f) \), where \( \alpha = 0.5 \text{ dB/cm/MHz} \)
- Max depth, example: \( d=23.1 \text{ cm}, c=1540 \text{ m/s} \)
- Time to travel up and down: \( T = 2d/c = 0.3 \text{ ms} \)
- Pulse Repetition Frequency PRF = \( 1/T = 3333 \text{ Hz} \)
- PRF = Framerate for A- and M-modes
- Too high PRF => Spatial ”aliasing”:

\[ \text{Tx pulse 1} \quad 23.1 \text{ cm} \quad \text{Tx pulse 2} \]
\[ \text{Time/2-way range, pulse 1} \quad \text{Time/2-way range, pulse 2} \]
\[ \text{Strong echo @ 25 cm} \quad \text{Spatial ”alias” @ 1.9 cm} \]
Modeling of ultrasound propagation

- Lossless wave equation, $u$ is displacement (or pressure):
  \[
  \nabla^2 u - \frac{1}{c_0^2} \frac{\partial^2 u}{\partial t^2} = 0
  \]

- Standard viscous wave equation – attenuation prop. to $\omega^2$
  \[
  \nabla^2 u - \frac{1}{c_0^2} \frac{\partial^2 u}{\partial t^2} + \tau \frac{\partial}{\partial t}(\nabla^2 u) = 0
  \]

  \[
  \nabla^2 u - \frac{1}{c_0^2} \frac{\partial^2 u}{\partial t^2} + \tau^\alpha \frac{\partial^\alpha}{\partial t^\alpha}(\nabla^2 u) = 0
  \]

- Derivatives of order $\alpha$ – not an integer - fractional derivatives:
  [Link to blog article](http://blogg.uio.no/mn/ifi/innovasjonsteknologi/content/bedre-diagnose-ved-bedre-derivasjon)
Resolution vs penetration

- Radial and lateral resolution improve with frequency
- Penetration falls with frequency
- Results in a near optimum frequency for an organ of a given size and depth:
  - 2.5 – 3.5 MHz cardiology
  - 5 – 7.5 MHz cardiology children; peripheral vessels
  - 3.5 – 5 MHz fetal imaging
- Pavlin, Foster, Ultrasound Microscopy 1998

**Figure 2.** Plot of depth of penetration versus frequency for a system with a dynamic range of 80 dB.
1D probe for 2D imaging

- 32-128 elements
- Beamforming in the azimuth plane for steering and control of the beam
- Acoustic lens for focusing in the elevation dimension

- Azimuth = in-plane = x
- Elevation = out-of-plane = y
Resolution

Two dimensions:
1. Radial or depth resolution
2. Azimuth or lateral resolution – orthogonal to beam direction.
   - Usually hardest to improve, as illustrated

Objekt Bilde
Object Image

III.: B. Angelsen, NTNU
Radial resolution (as in sonar ...)

- Resolution = half the pulse length
  \[ \Delta r = \frac{c \tau}{2} = \frac{c}{2B} \]

- Small value = good resolution = the ability to resolve two neighboring objects
- Also inverse proportional to bandwidth
- As bandwidth usually is a fraction of the center frequency, resolution is inverse proportional to center frequency, e.g. B=50% of \( f_0 \)
- Example \( f_0 = 3.5 \text{ MHz}, B = 0.5 \cdot 3.5 \text{ MHz} \Rightarrow \Delta r = \frac{1500}{(2 \cdot 1.75 \times 10^6)} \approx 0.43 \text{ mm} \)
Pulse compression in ultrasound?

**NO**
- Blind zone equal to the pulse length
- Depth-dependent filtering due to frequency dependent attenuation

**YES**
- Pulse Inversion in nonlinear acoustics
Lateral resolution

- Angular resolution for a source of size D:
  \[ \Theta_F \approx \frac{\lambda}{D} \]
- At a distance F (small angles approx.):
  \[ D_F \approx \Theta_F \cdot F = \frac{\lambda F}{D} = \frac{c F}{(f_0 D)} \]
- Res. best near the source (small F)
- Res. increases with frequency, \( f_0 \)
- Ex: \( f_0 = 3.5 \) MHz, \( \lambda = \frac{c}{f_0} = 1500/3.5 \times 10^6 \approx 0.43 \) mm
- Cardiology: \( D = 19 \) mm, \( \Theta_F = 0.43/19 = 0.023 \) rad = 1.3°. At depth 10 cm
  \[ D_F = 0.023 \times 100 = 2.3 \) mm
- Abdominal: \( D = 40 \) mm => \( \Theta_F = 0.62° \)
- Radial res. usually much better than lateral res. \( D_F = 2.3 \) mm >> \( \Delta r = 0.43 \) mm
Nonlinear pulse shape measured in water tank in our lab

Fabrice Prieur, Sept. 2009
Non-linear acoustics

The velocity of sound, $c$, varies with the amplitude, $s$:

$$\frac{dx}{dt} = c(t) = c_0 + (1 + \frac{B}{2A})s(t)$$

- $A$ and $B$ are the 1. and 2. order Taylor series coefficients for the pressure. $B/A$ is a measure of the non-linearity.
- $s(t) = \text{pressure} = p_0 + p_1(t)$
  - $p_0 = 1$ atmosphere
  - $p_1(t) = \text{applied pressure variation} (= \text{”signal”})$
- Two sources of nonlinearity:
  - Inherent in the material’s properties (equation of state): $B/2A$
  - Due to convection: the ’1’, exists even if material nonlinearity, $B/2A = 0$
Non-linear acoustics

• Positive peaks propagate faster than negative peaks:
  – Waveform is distorted.
  – More and more energy is transferred to higher harmonics as the wave propagates.
  – Eventually a shock wave is formed.

• B/A:
  – Linear medium: B/A = 0
  – Salt water: B/A=5.2,
  – Blood and tissue: B/A=6,..., 10.
Non-linear acoustics

• Positive effect on images:
  – 2. harmonic beam is narrower => better resolution
  – Is not generated in sidelobes of 1. harmonic beam => less sidelobes
  – Is generated inside medium => avoids some of the aberrations and reverberations from chest wall

• Negative effect:
  – 2. harmonic attenuates faster => less penetration

Whittingham, 2007
Liver

Fundamental 2. harmonic
Nonlinear acoustics, wave equations

\[ \nabla^2 p - \frac{1}{c_0^2} \frac{\partial^2 p}{\partial t^2} + \frac{\delta}{c_0^4} \frac{\partial^3 p}{\partial t^3} = -\frac{\beta}{\rho_0 c_0^4} \frac{\partial^2 p^2}{\partial t^2} \]

- Westervelt equation: \( p \) is pressure, \( c_0 \) is speed of sound, \( \rho_0 \) is density, \( \delta \) is loss factor, \( \beta = 1+B/A \) is nonlinearity coefficient
- Viscous loss, attenuation prop to \( \omega^2 \), good for water and air, not so good for medical ultrasound
- Prieur, Holm, ”Nonlinear acoustics with fractional loss operators,” Journ Acoust Soc Am, 2011, 2012 - other powers than 2
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Medical Ultrasound Scanner

• Imaging modes:
  – A-mode (A = amplitude)
  – M-mode (M = motion)
  – B-mode 2D (B = brightness)
    » Harmonic imaging – octave mode

• Doppler modes:
  – Doppler spectrum
  – Color doppler
  – (Strain)
A-scan (obsolete)

- **Figure 1** The pulse-echo principle is used to produce an ultrasound A-scan.
- A pulse is emitted from the transducer at the same time as a dot is set in motion from left to right on the A-scan screen.
- When an echo reaches the transducer, the received signal causes a vertical deflection of the trace.
- The distance between deflections on the A-scan corresponds to the depth of the interface from the transducer.

Peter N Burns: INTRODUCTION TO THE PHYSICAL PRINCIPLES OF ULTRASOUND IMAGING AND DOPPLER, November 2005
M-mode (M=motion) $\approx$ echo sounder
Heart
M-mode
Pulse Repetition Frequency

- \( d = \text{Attenuation}/(2 \cdot \alpha \cdot f) \), where \( \alpha = 0.5 \text{ dB/cm/MHz} \)
- Max depth, example: \( d = 23.1 \text{ cm}, c = 1540 \text{ m/s} \)
- Time to travel up and down: \( T = 2d/c = 0.3 \text{ ms} \)
- Pulse Repetition Frequency PRF = \( 1/T = 3333 \text{ Hz} \)
- PRF = Framerate for A- and M-modes
- Too high PRF \( \Rightarrow \) Spatial ”aliasing”:

\[ \text{Tx pulse 1} \]
\[ \downarrow \]
\[ 23.1 \text{ cm} \]
\[ \downarrow \]
\[ \text{Strong echo @ 25 cm} \]
\[ \text{Time/2-way range, pulse 1} \]
\[ \text{Tx pulse 2} \]
\[ \downarrow \]
\[ \text{Time/2-way range, pulse 2} \]
\[ \text{Spatial ”alias” @ 1.9 cm} \]
Linear Scan - Transmit Beamforming

Multiple beams are sent to scan the entire volume.
Frame rate in 2D

• From before: A-scan:
  – Max depth $d=23.1 \text{ cm}$, $PRF = \frac{c}{2d}= 3333 \text{ Hz}$

• Typical beamwidth (cardiac probe):
  $\theta = \frac{\lambda}{d} = 0.5\text{ mm}/19\text{ mm} = 0.026 \text{ rad} = 1.5^\circ$

• Desired sector size: $\Theta = 90^\circ$

• Beam distance ($p=25-50\%$ of beamwidth): $p \cdot 1.5^\circ = 0.5^\circ$

• No of beams per image: $N = \frac{\Theta}{p \cdot \theta} = \frac{90^\circ}{0.5^\circ} = 180$

• Frame rate: $FR = \frac{PRF}{N} = \frac{3333}{180} = 18.5 \text{ Hz}, \text{ fps}$
  – Analog TV: 50 half-frames per second
Parallel beamforming – multiple line acquisition

• Principle: use wider transmit beam and several, M, parallel receive beamformers
• Previous ex. with M=4:
  – $N = \Theta / (M \cdot p \cdot \theta) = 90^\circ / 2^\circ = 45$
  – FR = 3333/45 = 74 Hz, fps

2D framerate

- Assume a fixed aperture, D; a fixed maximum attenuation before next ping, A; and that attenuation varies linearly with frequency
  - PRF = 1/T = c/2d.
  - d = A/(2·α·f), where α = 0.5 dB/cm/MHz, and A is max attenuation => PRF = c·α·f/A
- Angular distance between receiver beams: \( \theta = p\lambda/D \), where p is the percentage shift relative to a beamwidth (p = 25-50% typ)
  - M = no of parallel rx beams, M= \( \theta_{tx}/\theta \), ratio of tx and rx beamwidths
- Angular sector to be covered: \( \Theta \)
- Number of beams: N = \( \Theta/\theta_{tx} = \Theta/M\Theta = \Theta D/p\lambda M \)
- FR = PRF/N = c\alpha fp\lambda M/(A\Theta D) = pM\alpha c^2/(A \Theta D)
  - Independent of frequency with these assumptions
  - Explanation: As frequency increases, frame rate decreases due to narrower beams. This cancels the increase in PRF due to a smaller depth.
• Doppler
• 3D
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G. York, Y. Kim, Ultrasound Processing and Computing: Review and Future Directions
Medical Ultrasound Markets

- Cardiology (Heart) 25% of market
  - Field where Norwegian company Vingmed was established in the 80’s as a startup from NTNU. Now GE Vingmed Ultrasound, center of excellence for cardiology ultrasound in GE Healthcare
  - Demo later in course
- Radiology (Inner organs) 40%
- Obstetrics/Gynecology 20%
- Niches
  - Vascular
  - Urology (B&K Medical, Denmark)
  - Surgery (Medistim AS and Sonowand AS)
  - Emergency medicine (driver for handheld ultrasound)
  - General practice
  - Dermatology (skin)
  - Veterinary
  - Small animals for medical experiments/testing
  - ...
High-end Ultrasound providers

• Philips
  – Acquired ATL, Seattle WA, and HP/Agilent, Boston MA

• Siemens
  – Acquired Acuson, Silicon Valley

• GE Healthcare
  – Acquired Vingmed

• Japan: Toshiba, Hitachi, Aloka

• Emerging? France: Supersonic Imagine
www.geultrasound.com
Laptop ultrasound

GE Logiq Book XP

Sonosite M-Turbo
Pocket Ultrasound: GE VScan

- Developed by GE Vingmed Ultrasound in Norway
  - For cardiologists or primary care
  - Liver, gall bladder, kidney, fetal position, cardiac, aorta
- No 14, Time Magazine’s list The 50 Best Inventions of 2009
- Winner Årets ingeniørbragd, Norway 2009
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Sector scan

- Small footprint:
  - Access through ribs for cardiology
  - Typ: 19 mm adult, <12 mm pediatry
- Started with mechanically tilted probes (figure from B. Angelsen, NTNU)
- Now: phased multi-element arrays and electronic tilting by means of digital delays
Linear scan

- Beam is moved along probe surface by switching elements in and out
- Can be done with simple electronics (important in the 80’s)
- Suitable for organs with good access, e.g. external organs:
  - Artery in neck (halspulsåre)
  - Thyroid (Skjoldbrukskjertel)
  - Muscles, tendons
Curvilinear scan

- Beam is scanned along probe surface like linear scan
- Results in a large sector imaged with moderate requirements on access
- Used for
  - Fetal imaging
  - Internal organs: kidney, liver
  - Breast ...

III.: B. Angelsen, NTNU
Factors which determine frame rate

- Depth where attenuation reaches e.g. ~60 dB
  - Attenuation = 2d·\(\alpha\)·f
  - Example: 60 dB attenuation (\(\alpha = 0.5\) dB/cm/MHz):
    » f = 2.6 MHz: d = 23.1 cm depth
- Speed of sound
- Number of transmit beams in M-, 2-D, or 3-D modes
- Recap 1-D and 2-D framerate on next slides
Pulse Repetition Frequency

- Max depth, example: $d=23.1\ \text{cm}$, $c=1540\ \text{m/s}$

- Time to travel up and down: $T = \frac{2d}{c} = 0.3\ \text{ms}$

- Pulse Repetition Frequency $\text{PRF} = \frac{1}{T} = 3333\ \text{Hz}$

- $\text{PRF} = \text{Framerate for A- and M-modes}$
Frame rate in 2D

• Previous A-scan:
  – Max depth d=23.1 cm, PRF = c/(2d)= 3333 Hz
• Typical beam width (cardiac probe): \( \frac{\lambda}{d} = \frac{0.5\text{mm}}{19\text{mm}} = 0.026 \text{ rad} = 1.5^\circ \)
• Beam distance (typ 25-50% of bw, here 1/3): 0.5°
• Desired sector size: 90°
• Number of tx beams per image: \( N = \frac{90^\circ}{0.5^\circ} = 180 \)
  – Frame rate: \( FR = \frac{PRF}{N} = \frac{3333}{180} = 18.5 \text{ Hz or fps} \)
• No of parallel receive beams: \( M=4 \Rightarrow N=180/4=45 \)
  – Frame rate \( FR = \frac{PRF}{N} = \frac{3333}{45} = 74 \text{ fps (frames per second)} \)
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Bioeffects and Safety

- Thermal effects: heating of muscles, tendons (physiotherapy)
- Mechanical effects: crushing of kidney/gall stones, surgery
- All instruments display TI and MI
- TI – Thermal Index: estimate of heating in hottest part of beam
  - TIS - Thermal Index Soft tissue: the most usual one
  - TIB - Thermal Index Bone: for bone in focal point (fetal scan)
  - TIC - Thermal Index Cranial: bone at probe surface (cranial imaging)
  - TI <1.5 centigrade is considered safe regardless of exposure time
- MI - Mechanical Index. MI = \( p/f^{0.5} < 1.9 \)
  - Risk of cavitation (sometimes called cavitation index)
  - Peak negative pressure in MPa
  - Frequency in MHz
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Emerging technology

• 3D ultrasound:
  – Not really emerging, but still quite new
• Elastography
• Contrast agents
  – Targeted ultrasound imaging
  – Ultrasound-assisted drug delivery
• Ultrasound tomography
Tissue: about 70% water

Compression

Shear

Probing of the mechanical structural properties of the SOLID components requires the application of SHEAR deformation.
Palpation and Elasticity in human soft tissues

Young’s Modulus $E$

$$E = \frac{\sigma}{\varepsilon} = \mu \frac{3\lambda + 2\mu}{\lambda + \mu}$$

2 other mechanical coefficients are commonly used for defining the elasticity of a solid material:

$\lambda \sim$ Bulk Compression Modulus, almost constant, of the order of $10^9$ Pa,

$\mu$ Shear Modulus, Strongly heterogeneous, varying between $10^2$ and $10^7$ Pa

$\lambda \gg \mu \Rightarrow E \approx 3\mu$

Mickael Tanter
Mechanical waves in soft tissues

Compressional Waves propagate at
\[ c_p = \sqrt{\frac{\lambda + 2\mu}{\rho}} \approx \sqrt{\frac{\lambda}{\rho}} \approx 1500 \text{ m.s}^{-1} \]

Shear waves propagate at
\[ c_s = \sqrt{\frac{\mu}{\rho}} \approx 1-10 \text{ m.s}^{-1} \]

Two kind of waves propagating with totally different speeds !!

Shear waves propagate only at low frequencies < 1000 Hz
Ultrasound-based methods

- Static
  - Not covered here
- Transient
- Acoustic streaming = acoustic radiation force
The Transient Elastography Technique

Shear wave generation + Ultrafast Imaging

Low Frequency « punch »
(50 Hz, 2 cycles)

~ 20 ms

Acquisition of RF signals
stored
In memories
*(ultrasound pulsed excitation at 4 MHz)*

~ 1 ms

100 ms

Time
Fibroscan: Echosens, Paris

ULTRASOUND ELASTOGRAPHY
A major contribution to medical practices

The FibroScan® technique is used to quantify hepatic fibrosis in a totally non-invasive and painless manner, with no contra-indications for the patient.
Remote Palpation using the Ultrasonic Radiation force

\[ F(\vec{r}, t) = \frac{\alpha}{\rho c^2} \left(\vec{p}(\vec{r}, t)\right) \]

\( \alpha \) is absorption coefficient
Breast Cancer Examples

2ème lésion ACR4 non vue à la mammographie.
Taille : 2mm.

FPS : 5.4 – Arbitration : cineloop – max echo id : 308/1509, seis : 0 – filter deactivated new persistence 0.720159

Investigational Use Only – Not for Diagnosis

E > 150 kPa

2 carcinomes canalaire Invasifs (grade I & RH+}

Mickael Tanter
In vivo assessment of carotid plaque elasticity

150 µm resolution in the elasticity image

Mickael Tanter
SH, 72
Elastography

- Shear waves probe elasticity and correlate better with palpation than conventional ultrasound
- Shear wave described by:
  - Young’s modulus $E \approx 3 \mu$
  - Shear modulus $\mu$, complex shear modulus $G^* = G_d + i G_l$
  - Speed of propagation $c^2 = \mu/\rho$
- Ultrasound based elastography:
  - Static, dynamic, acoustic streaming (supersonic)
- MR-based elastography
- Hypothesis: Fractional strain-stress relationship can explain power law behavior of shear waves
Literature

- Wells, Peter NT, Hai-Dong Liang, and Terry P. Young. "Ultrasonic imaging technologies in perspective." Journal of Medical Engineering & Technology 35.6-7 (2011): 289-299
- Sverre Holm, *Medisinsk ultralydavbildning*, 2008