Ultrasonic Radiation
What Is It?

• Longitudinal **pressure** waves
• Requires a medium for transmission
Why Do We Care?

• Some potential for damage due to tissue heating, mechanical stress/strain
Why Do We Care?

• Regulatory
  – REDA for therapeutic ultrasound
  – Safety Code 24 for industrial (“low frequency”) ultrasound
Physical Principles
Ultrasound

• First postulated in 1794 by Spallanzini to explain bat navigation
• Sound “spectrum”

<table>
<thead>
<tr>
<th>Audible</th>
<th>US cleaners</th>
<th>Medical US</th>
</tr>
</thead>
<tbody>
<tr>
<td>20</td>
<td>20k</td>
<td>1M</td>
</tr>
<tr>
<td>50k</td>
<td></td>
<td>50M</td>
</tr>
</tbody>
</table>
Simple Harmonic Motion

- Sound waves are longitudinal
Modeling Sound Waves

• Wave equation for pressure

\[
\frac{\partial^2 p}{\partial z^2} = \left( \frac{\rho}{\kappa} \right) \frac{\partial^2 p}{\partial t^2} = \frac{1}{c^2} \frac{\partial^2 p}{\partial t^2}
\]

• \( \rho \): density, \( \kappa \): bulk modulus of elasticity

\[
\kappa = -V \frac{dp}{dV}
\]
Modeling Sound Waves

• Relationship between pressure and particle velocity

\[- \frac{\partial p}{\partial z} = \rho \frac{\partial u}{\partial t}\]

• Pressure and particle acceleration

\[a = -\frac{1}{\rho} \nabla p\]
Simple Harmonic Motion

• Plane wave approximation

\[ p(z, t) = Ae^{i(kz - \omega t)} \]

\[ k = \frac{2\pi}{\lambda} \]

\[ \omega = 2\pi f \]

• Valid for many medical applications
Derived Properties

• (Specific) Acoustic Impedance

\[ Z = \frac{p}{u} = \rho c \quad [kg \cdot m^{-2} \cdot s^{-1}] \]

• Used to characterize a medium’s resistance to being disturbed by sound waves
• Generally location dependent
• Analogous to “index of refraction” in optics
Derived Properties

• Sound Intensity (W/m²)
  – Energy/volume * speed of wave
  – Instantaneous *versus* time averaged
  – Most radiation protection quantities are related to sound intensity or sound pressure level
  – Related quantities – intensity derived from sound pressure
Sound Intensity

• Time-averaged

\[ I(r, t) = \frac{1}{\tau} \int_t^{t+\tau} p(r, t') u(r, t') \]

\[ = \frac{1}{2} p_{\text{max}} u_{\text{max}} \]
Sound Intensity

• In terms of particle velocity

\[ I = \left(\frac{\rho u_{\text{max}}^2}{c}\right) \]

• Pressure

\[ I = \frac{p_{\text{max}}^2}{2\rho c} \]

• Decibel scale (relative to some reference intensity or pressure, 1-10 MPa \( p_o \) for diagnostic)

\[ dB = 10 \log\left(\frac{I}{I_o}\right) = 20 \log\left(\frac{p}{p_o}\right) \]
Ultrasonic Sources
Ultrasound Transducer

- Used to generate and detect ultrasound by piezoelectric effect (discovered 1880)
Modern Piezoelectric Transducer

Shapes beam (later)
Piezoelectric Effect

- Crystal deforms when a voltage is put across it – conversion of electrical into mechanical energy
- High frequency alternating current causes the crystal to vibrate at that frequency
  - Crystal continues to “ring” at some resonant frequency; resonance can also be used for pulsed voltages
  - Crystal produces a voltage when under strain – detection of ultrasound
Piezoelectric Materials

- Choice is typically either: lead-zirconate-titanate (PZT) ceramic, or poly-vinylidene-difluoride (PVDF) polymer film

<table>
<thead>
<tr>
<th></th>
<th>PZT ceramics</th>
<th>PVDF films</th>
</tr>
</thead>
<tbody>
<tr>
<td>Acoustic impedance</td>
<td>High, requires matching layers for efficient transfer of acoustic energy to sample</td>
<td>Low, well matched to water</td>
</tr>
<tr>
<td>Malleability</td>
<td>Low (porous solid)</td>
<td>High (elastic)</td>
</tr>
<tr>
<td>Electrical impedance</td>
<td>Low, can be efficiently matched to electronics</td>
<td>High, can be difficult to match to electronics</td>
</tr>
<tr>
<td>Sensitivity</td>
<td>Higher</td>
<td>Lower</td>
</tr>
</tbody>
</table>
Piezoelectric Materials

• No obvious preference
  – PZT more typical because of sensitivity (dominant for applications up to 25 MHz)
  – Resonant frequency depends on thickness, $\lambda_{\text{res}} = 2 \cdot \text{crystal thickness}$ ($\lambda_{\text{res}}$ measured in the crystal)

• Pulse length from transducer characterized by “Q factor” – very important quantity for applications
Individual Piezoelectric Elements In An Array Transducer

Formed from a single crystal and acoustic matching layers
Transducer Q-Factor

• Backing material in transducer used to reduce ring-down time (decrease Q) to absorb back-emitted vibrations
• Shorter ring-down time means shorter pulse length, superior **axial resolution**
• High Q transducers for Doppler measurements – more pure frequency for measuring frequency Doppler shift
Transducer Q-Factor

![Graph showing Transducer Q-Factor with two curves: Curve A (Q=20) and Curve B (Q=2). The graph plots Relative Amplitude against Relative Frequency. The term "Resonance Frequency" is indicated.]

Ring down-time of a high Q and low Q transducer.
Axial Resolution

[Diagram showing transducer, resolved structures, and axial distance with pulse, resolved echoes, and not resolved regions]
Focused Ultrasound Fields

- Focal depth: \( R \); Depth of field: \( \text{DOF} \)
- Based on diffraction theory, **lateral resolution** related to FWHM for a spherically focused array

\[
\text{FWHM} = \frac{c \cdot R}{v_o \cdot d} = \lambda \cdot f \# 
\]
Focused Ultrasound Fields

• What is $f#$?
  – Ratio: distance to focal plane / diameter of radiating surface
Focused Ultrasound Fields

- Lateral resolution can also be improved by reducing $f#$ (bigger, more curvature)
- Depth of field over which beam in focus also decreases

\[ \text{DOF} \propto \lambda \cdot (f\#)^2 \]

- Optimal $f#$ chosen to maximize lateral resolution while achieving required depth of field for the procedure
Lateral Resolution

LATERAL SCANNING DIRECTION

AXIAL DIRECTION

BEAM WIDTH

RESOLVED STRUCTURES

NOT RESOLVED
Depth Of Field

Resolution is poorer out of the focal plane for low f# transducer
Compromising On $f#$
Electronic Focusing

• Depth of field issues can be partially overcome by electronic focusing, delaying individual transducer elements by different times in different sections of the array (later)

• Most modern transducers use this technology
Lateral Resolution And Wavelength (Frequency)

- Lateral resolution improves with increased frequency
- Penetration decreases

![Graphs showing the relationship between frequency and lateral resolution or penetration.](image-url)
Interactions with Tissue
Interaction Types

• Absorption
• Scattering
• Reflection and Refraction
Absorption

• Removal of useful ultrasound energy and conversion into heat
  – Viscous losses associated with molecular oscillations
  – Heat conduction from regions of high pressure to rarefaction
  – Induction of transient changes in molecular structure, vibrational and/or rotational states
  – Increases with frequency ($f^{1-2}$ depending on tissue)
Scattering

- Ultrasound waves get scattered in all directions by tissue heterogeneities (cross sectional area $a$)
- Results in distinct “speckle” pattern on US images
- Type of scattering depends on scatterer dimension compared to wavelength
## Scattering

<table>
<thead>
<tr>
<th>Scale of Interaction</th>
<th>Frequency Dependence</th>
<th>Scattering Strength</th>
<th>Examples</th>
</tr>
</thead>
<tbody>
<tr>
<td>( a &gt; \lambda ): Geometric region, ray theory for reflection/refraction</td>
<td>( f^0 )</td>
<td>Strong</td>
<td>Diaphragm, large vessels, soft tissue / bone, eye orbit</td>
</tr>
<tr>
<td>( a \sim \lambda ): Stochastic region (diffractive)</td>
<td>Variable</td>
<td>Moderate</td>
<td>Predominates for most structures</td>
</tr>
<tr>
<td>( a &lt; \lambda ): Rayleigh region</td>
<td>( f^4 )</td>
<td>Weak</td>
<td>Red blood cells</td>
</tr>
</tbody>
</table>
Reflection

• Occurs at boundaries where there is an acoustic impedance mismatch
  – Sudden change in wave speed
  – Applicable for target sizes $a \gg \lambda$
  – Analogous to optical radiation

• Diagnostic information obtained from reflections
Reflection

• Fraction reflected depends on difference in acoustic impedance and angle of incidence:

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

• At normal incidence:

\[ R = \left( \frac{Z_2 - Z_1}{Z_2 + Z_1} \right)^2 \]
Acoustic Impedance

- A property of the medium, \( Z = \rho c \)

<table>
<thead>
<tr>
<th>Material</th>
<th>( Z \times 10^6 \text{ kg/m}^2 \text{ s} )</th>
<th>( c ) (m/s)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Air</td>
<td>0.0004</td>
<td>331</td>
</tr>
<tr>
<td>Fat</td>
<td>1.38</td>
<td>1450</td>
</tr>
<tr>
<td><strong>Water</strong> (50 °C)</td>
<td>1.54</td>
<td><strong>1540</strong></td>
</tr>
<tr>
<td>Brain</td>
<td>1.58</td>
<td>1541</td>
</tr>
<tr>
<td>Blood</td>
<td>1.61</td>
<td>1570</td>
</tr>
<tr>
<td>Kidney</td>
<td>1.62</td>
<td>1561</td>
</tr>
<tr>
<td>Liver</td>
<td>1.65</td>
<td>1549</td>
</tr>
<tr>
<td>Muscle</td>
<td>1.70</td>
<td>1585</td>
</tr>
<tr>
<td>Skull (bone)</td>
<td>7.8</td>
<td>4080</td>
</tr>
<tr>
<td>Piezo. polymers</td>
<td>4.0</td>
<td>2300</td>
</tr>
<tr>
<td>PZT-4</td>
<td>30.0</td>
<td>4000</td>
</tr>
</tbody>
</table>
Reflection

• Example: air to water (such as transducer to tissue)

\[ R = \left( \frac{Z_2 - Z_1}{Z_2 + Z_1} \right)^2 \]

\[ = \left( \frac{1.54 - 0.0004}{1.54 + 0.0004} \right)^2 \]

\[ \approx 1 \]

• Nothing gets into tissue – use an impedance matching gel!
Reflection
Reflection

• Example: muscle to bone

\[ R = \left( \frac{Z_2 - Z_1}{Z_2 + Z_1} \right)^2 \]

\[ = \left( \frac{7.8 - 1.70}{7.8 + 1.70} \right)^2 \]

\[ \approx 0.41 \]

• Strong signal from bone surface, but can’t see much behind bone
Reflection

muscle

bone

usra.ca
Image Formation

Diagram:

- Transducer
- Ultrasound Pulse
- Eye
- Time (μs)
- "A Scan"
- "B Scan"
- Video Monitor
Refraction

- Sound waves crossing a boundary obey Snell’s Law

\[
\frac{\sin \theta_i}{\sin \theta_t} = \frac{c_1}{c_2} \quad \text{and} \quad \theta_i = \theta_t
\]
Overall Attenuation (Intensity)

- Attenuation from ALL sources
- Rule of thumb for most soft tissues: $\alpha \approx 1 \text{ dB cm}^{-1} \text{ MHz}^{-1}$

Conversion to linear attenuation for SAR:

$$\frac{I}{I_o} = e^{-\mu x}$$

$$dB = 10 \log \left( \frac{I}{I_o} \right)$$

$$e^{-\mu_a z} = 10^{-\frac{\alpha}{10}}$$

$$\mu_a = 2.303 \left( \frac{\alpha}{10} \right)$$
Overall Attenuation

• More often, the attenuation is given in terms of the pressure wave

• In that case: \( \frac{P}{P_o} = e^{-\mu p x} = e^{-\frac{\mu}{2} x} \)

• Factor of 2 represents the \( p^2 \) dependence of intensity

• In this formulation our rule of thumb becomes

\[ \alpha_p \approx 0.5 \text{ dB cm}^{-1} \text{ MHz}^{-1} \]

  • Culjat et al. (2010) on the website uses pressure
Biological Effects
Thermal Effects

• Heat generation

\[ SAR = I \frac{\mu_a}{\rho} \left[ \begin{array}{c} Power \\ kg \end{array} \right] \]

• Evolution of heating with time (approx)

\[ \rho c_p \frac{\partial T}{\partial t} = \mu_a I + \kappa_{th} \nabla^2 T \]

At early time \( t \):

\[ T = \left( \frac{\mu_a I}{\rho c_p} \right) t + T_0 \]

Steady-state:

\[ -\mu_a I = \kappa_{th} \nabla^2 T \]
Bioheat Transfer Equation

\[ c \frac{\partial T}{\partial t} = SAR + P_{\text{metabolic}} - P_{\text{conduction}} - P_{\text{convection}} \]

\[ SAR = I \left[ \frac{w}{cm^2} \right] \frac{\mu}{\rho} \]

\[ P_{\text{cond}} = -\nabla \left( k \nabla T / \rho \right) \]

\[ P_{\text{conv}} = \rho_b c_b F (T - T_b) \]

\[ \frac{\partial T}{\partial t} = \frac{SAR}{c} \quad * \text{Worst case for heating} * \]
Bioheat Transfer Equation

- Putting it all together...

\[ \rho c \frac{\partial T}{\partial t} = \mu_a I + \nabla (k \nabla T) - \rho_b c_b F (T - T_b) \]
What Does It Mean?

• Determining temperature distributions in tissue is extremely complicated
Experimentally Determined Distributions

• Heating is greatest near bone
  – High absorption in bone, which then acts like a radiator, heating surrounding soft tissues (e.g. neonatal brain)
  – May define pain threshold for an individual

• Maximum heating of a fetus during standard diagnostic US estimated to be less than 2°C for SPTA intensity < 720 mW/cm² (later)
  – Heating of 5+°C is possible when bone present later during pregnancy
Heating

• **Australian Ultrasound Society**
  – > 1.5°C not harmful over an extended time
  – Fetal temperature > 41°C for > 5mins may be harmful

• **American Institute of Ultrasound in Medicine**
  – Exposures up to 50hrs, no biological effects with 2°C elevation
  – No effects observed for \( \Delta T \leq \frac{6 - \log t}{0.6} \)
Non-Inertial (Stable) Cavitation

• Bubble expansion and contraction
  – Bubbles of gas in tissue oscillate with the sound wave
  – Not well understood, but not a concern

• Strong forces at the periphery of the bubble
Inertial (Transient) Cavitation

• “Cavitation collapse”
  – Bubbles grow until adiabatic contraction
  – Results in high local temperatures and pressures – microscopic bubbles produced by the sound waves compress, producing shock waves

• Sonoluminescence
  – Need lots of energy; light emission used to estimate temperatures up to 10000 K

• Free radical production?
Sonoluminescence

- Increases with frequency, but not observed above 2 MHz
- Appears at a threshold intensity and increases linearly; may disappear at too high intensity
- Decreases with increased ambient temperature and pressure
Radiation Force

- Unbalanced net force due to ultrasound
- Can cause red blood cells to bind together, impeding flow
- May also cause shear stresses at fluid boundaries
Power Levels

• All effects strongly dependent on power density (W/cm$^2$)
  – Diagnostic procedures typically < 100 mW/cm$^2$
  – Physiotherapy/therapeutic heating ~1 W/cm$^2$
  – Intentional thermal destruction >10 W/cm$^2$
# Power Levels *In Vivo*

<table>
<thead>
<tr>
<th>Mechanism</th>
<th>Freq. (MHz)</th>
<th>I (W/cm²)</th>
<th>t (min)</th>
<th>Model system</th>
<th>Effect</th>
</tr>
</thead>
<tbody>
<tr>
<td>Thermal</td>
<td>0.9</td>
<td>5</td>
<td>10</td>
<td>Dog femur</td>
<td>Marrow damage</td>
</tr>
<tr>
<td>Thermal</td>
<td>1.0</td>
<td>1</td>
<td>5</td>
<td>Pregnant mouse</td>
<td>Fetal wt. reduction</td>
</tr>
<tr>
<td>Thermal</td>
<td>2</td>
<td>1</td>
<td>5</td>
<td>Pregnant mouse</td>
<td>Fetal wt. reduction</td>
</tr>
<tr>
<td>Radiation force</td>
<td>3</td>
<td>1</td>
<td>8x10⁻⁴</td>
<td>Chick embryo</td>
<td>Blood flow stasis</td>
</tr>
<tr>
<td>Stable cavitation</td>
<td>0.75</td>
<td>0.3</td>
<td>2</td>
<td>Guinea pig leg</td>
<td>Bubbles observed</td>
</tr>
</tbody>
</table>
Medical Applications
Diagnostic Usage

Success of ultrasound imaging due to:

1. High soft tissue contrast
2. Rapid image formation (real-time imaging is possible)
3. Inexpensive
4. Safety, especially compared to ionizing radiation
A-Scan

- “Amplitude” scanning mode
- Low Q transducer
- Obsolete
- Uses time-of-flight of echoes to locate boundaries in tissue
- Vastly improved upon by...
B-Scan

• “Brightness” scanning mode (low Q)
• Standard diagnostic imaging technique for real-time scanning (pulse-echo mode)
• Imaging performed by sending a pulse (2-3 $\lambda$ long) along each of many lines comprising a single frame, and processing the reflections
• Images formed on a video display, displaying each echo as a dot at the correct position on its line with brightness proportional to the echo’s amplitude
  – On screen brightness is corrected by assuming some average loss/depth and applying a gain to the signal
Pulse-Echo Imaging
Array Scanning

... Individual piezoelectric elements

US beam line #1

#2

#3
Array Scanning
Electronically steered/focused US beams by delaying pulses: sound waves interfere to form the wave front.
Scanning Image Formation

Initial position of US beamline

Direction of sweep of US beamline
Frame Rate

- “Flicker free” at approximately 16 frames per second (fps), ideally ~30 fps
- Example:
  - 225 transducer elements per frame
  - x16 frames = 3600 lines/sec
  - Thus, single “line” completed in 1/3600 = 278 µs
  - Corresponding round trip distance = 1540 m/s * 0.278 s = 42.8 cm
  - So you can image to a depth of ~21.4 cm
Frame Rate

• Example demonstrates compromises between:
  – Depth of imaging and frame rate
  – Lines per frame (# transducer elements) and frame rate
3D Ultrasound

• Enabled by real-time position and orientation tracking of a B-mode transducer
• Sophisticated computer software reconstructs a series of 2D scans to form a 3D image
• Resolution tends to be lower than individual B-mode scans, but additional diagnostic information may be available
  – *e.g.* deformations that may not be apparent in 2D
3D Ultrasound
Continuous Doppler Ultrasound

• Used for examining surface vessels and fetal heart sounds
Continuous Doppler Ultrasound

- Emitted pulse Doppler shifted twice
  - Moving structure sees a shifted frequency \( f' \), and pulse shifted again upon reflection

\[
f' = \frac{f}{1 - \frac{v}{c} \cos(\theta)} = f + \Delta f
\]

\[
\Delta f \approx f \left( \frac{v}{c} \right) \cos(\theta)
\]

So measured shift at transducer

\[
\Delta f \approx 2f \left( \frac{v}{c} \right) \cos(\theta)
\]
Doppler Example

- c=1540 m/s, f=5 MHz, \( \theta \)=45\(^\circ\), v=5 cm/s
  - \( \Delta f \approx 2 \times 5 \times 10^6 \times (0.05 / 1540) \times \cos(45) \)
  - \( \approx 230 \) Hz

- Note that this is in the audible range
  - Shift frequency is often played as an audio signal (especially fetal heart monitoring)
Continuous Doppler Ultrasound

- Limited by the complete absence of depth information (continuous mode)
- Signal is superposition of all moving structures in the beam
  - Most useful for isolated, superficial structures
- Performed using high Q transducer
Pulsed Doppler Ultrasound

• Limitations for continuous Doppler imaging can be alleviated using pulsed Doppler
  – Pulsed continuous versus colour flow imaging
• Frequency shift can then be related to a given depth based on time-of-flight
• Lower Q transducer (for pulse width), so broader frequency distribution and less sensitivity to small frequency shifts (colour flow imaging)
Doppler Colour Flow Imaging

• Combines Doppler imaging with real-time scanning in a non-trivial way
• Result is an image of Doppler shifts, colour-coded to represent magnitude AND direction of flow (i.e. velocity), either toward or away from transducer
Doppler Colour Flow Imaging
Ultrasound for Therapy

• **Physiotherapy**
  – Pain relief
  – Accelerating tissue regeneration
  – Stimulating capillary circulation

• Thermal and probably non-thermal mechanisms

• Pain threshold ~3-5 W/cm², start at 1 and increase to threshold

• 5-15 minutes, move transducer constantly to avoid hot spots
Ultrasound for Therapy

• Hyperthermia, thermal coagulation, surgery, dental scaling
• Concentrated, localized heating using focused ultrasound (1 – 10 MHz)
• Shaking / emulsification (18 – 40 kHz)
  — e.g. cataract emulsification to facilitate aspiration
Ultrasound for Therapy

• Extracorporeal Shockwave Lithotripsy (ESWL)
  – Intense ultrasound to pulverize kidney stones so that they can be passed normally without surgery
High Power US Generation

• Electrical discharge of a high voltage current across a spark gap in a water filled container
  – Creates a vaporization bubble in the water, resulting in a high intensity pressure wave
ESWL Delivery
ESWL Therapy

- Focus of the ultrasound source centred on kidney stone
- Coupling of sound wave to patient by water bath (patient submerged) or water-filled cushions
- Treatment of choice for over 80% of stones in kidney and ureter
High Intensity Focused Ultrasound

• “HIFU”

• Highly focused ultrasonic beams can also be produced using **phased arrays**

• These modern systems are often used to perform “ultrasonic surgery”, including lithotripsy, hyperthermia of tumours
HIFU Example
Ultrasonic Tomography

- CT using an ultrasound!
- Main application is mammography
- Modern systems use:
  - Reflected waves (structure)
  - Transmitted waves (attenuation)
  - Time-of-flight (speed of sound)
Principles of Ultrasonic Tomography

• Multiple 2D image slices through 3D object
• US waves transmitted through object and detected on far side
• Basic idea is “to determine the distribution of [the] objective function or acoustic speed and attenuation through the measurement of [the] scattering field at a known set of boundary sites” [Zhao 2005]
Methods for reconstruction

• Straight ray approximation
  – Lines of constant phase (assumption: unrefracted)
  – Used for transmission or reflection tomography

• Diffraction tomography
  – Helmholtz equation with
    • Born (backscattering) approximation
    • Rytov (forward propagation) approximation
  – Inversion of approximate wave equation
  – Finite element method or similar approach to solve
Development
In vivo human imaging
Colour coded images
Detection and Dosimetry
Measuring Ultrasound

• Direct
  – Measure sound pressure

• Indirect
  – Measure temperature increase

• Quality assurance (QA) in ultrasound is really the calibration of the measurement instrument
Direct Measurement

• Hydrophone (alone, or in array)
  – Small receiving piezoelectric transducer(s)
  – Measure acoustic pressure from the voltage generated in the crystal
    • Pressure gives time-averaged ultrasound intensity
      \[ I = \frac{p_{\text{max}}^2}{2\rho c} \]
  – Must not significantly perturb the US field
  – Low Q (~3) preferred – flat frequency response under 15 MHz
  – Ideal for field usage
Direct Measurement

• Radiation force balance
  – Used as a primary standard to calibrate hydrophones
  – Measures force directly on a flat object (a vane) – force converts to pressure
  – Theory is complex
Primary & Secondary Calibration Standards

• Primary standards are used to calibrate other (secondary) standards
  – Example, a radiation force balance at the NRC or NIST

• Secondary standards are used by institutions to calibrate field instruments
  – Example, you send a hydrophone to an accredited laboratory to be calibrated, and you calibrate your “field” instruments against it
Indirect Measurement

• Measure temperature increase using common temperature transducers
  – Thermocouples, thermistors, *et cetera*
Dosimetry

• “Dose” in ultrasound usually means
  – Temperature increase distribution, as this is the main biological effect of concern
  – Specific absorption rate distribution, where recall that

\[
SAR = I \left( \frac{\mu_a}{\rho} \right) \frac{W}{kg}
\]
Dosimetry

• SAR in ultrasound is analogous to dose rate (Gy/s) in ionizing radiation

• Interpreting SAR distribution as something biologically meaningful is difficult, because SAR influence is complicated
  – Rate of energy deposition does not directly correlate to temperature rise, due to conduction and convection
  – In reality, if we are concerned with thermal effects then temperature distribution is more useful than SAR
Protective Measures
Protective Measures

• None
• US from medical devices doesn’t get into the air, so no danger to staff
Exposure Standards
Safety Code 23

• Now known as “Guidelines for the safe use of ultrasound”
  – Safety Code 24 for industrial ultrasound

• Safety codes are released by Health Canada

• “At present, there are few standards in Canada related to ultrasound. The standards that exist are emission standards, not exposure standards.”
Diagnostic Ultrasound

• See updated (2001) “Guidelines for the Safe Use of Diagnostic Ultrasound”
• No exposure regulations at present, but recommendation that spatial-peak-temporal-average (SPTA) < 720 mW/cm$^2$
  – Fetal heart monitoring, < 20 mW/cm$^2$
  – Ophthalmic devices, < 50 mW/cm$^2$
• Prior to 2001 update, guidelines for fetal examination specify (SPTA) power < 100 mW/cm$^2$
SPTA Intensity

- Spatial-peak-temporal-average intensity is one of the main exposure quantities in ultrasound safety assessment.

\[ I_{SPTA} = \text{Max} \left\{ \frac{1}{\tau} \int I(\vec{r}, t) \, dt \right\} \frac{mW}{cm^2} \]

- Maximum peak intensity averaged over 1 second in a 1 cm\(^2\) area.

- Attempts to account for variations over time and space, important for pulsed beams.
Review of Biological Effects (Safe Use Guide)

1. A diagnostic ultrasound exposure that produces a maximum *in situ* temperature rise of no more than 1.5 °C above normal physiological levels (37 °C) may be used clinically without reservation on thermal grounds,

2. A diagnostic ultrasound exposure that elevates embryonic and fetal *in situ* temperature above 41 °C (4 °C above normal temperature) for 5 minutes or more should be considered potentially hazardous,

3. The risk of adverse effects is increased with the duration of exposure.

4. With regard to adult tissues, the available literature suggests that tissue temperature elevations in the range of 8-10 °C, sustained for 1 to 2 minutes will cause tissue injury.
Ultrasound Therapy

• Radiation Emitting Devices Act (REDA) regulation for SPTA power < 3 W/cm²
• Warning sign posted on therapeutic devices only
Ultrasound for Surgery

• Surgery, hyperthermia, lithotripsy (blowing up kidney stones), and scaling

• No existing recommendations in Canada since heat and destruction is intended