Projection Radiography

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Based on J. L. Prince and J. M. Links, Medical Imaging Signals and Systems, and lecture notes by Prince. Figures are from the textbook.
Lecture Outline

• Instrumentation
  – X-ray tube configuration
  – Filtration and restriction of x-ray photons
  – Compensation and Scatter control
  – Film screen detector

• Image formation
  – Geometric effect
  – Extended source
  – Detector/film response

• Image quality
  – Contrast and SNR
  – Effect of noise and Compton scattering
Overview

- **Systems:**
  - chest x-rays, mammography
  - dental x-rays
  - fluoroscopy, angiography

- **Properties**
  - high resolution
  - low dose
  - broad coverage
  - short exposure time
Radiographic System

Diagram of a radiographic system with labeled parts:
- x-ray tube and housing
- collimator
- patient
- table
- ionization chamber
- grid
- cassette
- Bucky assembly

Diagram shows the path of x-rays from the x-ray tube through the patient to the cassette on the Bucky assembly.
X-ray Tube

Rotating anode
Cathode

anode assembly

filament circuit

ground

high voltage

stator

x-rays

Cathode assembly
X-Ray Tube Components

- **Filament** controls tube current (mA)
- **Cathode** and **focussing cup**
- **Anode** is switched to high potential
  - 30–150 kVp
  - Made of tungsten
  - Bremsstrahlung is 1%
  - Heat is 99%
  - Spins at 3,200–3,600 rpm
- **Glass housing; vaccum**
Exposure Control

- kVp applied for short duration
  - fixed timer (SCR), or
  - automatic exposure control (AEC), 5 mm thick ionization chamber triggers SCR
- Tube current mA controlled by
  - filament current, and
  - kVp
- mA times exposure time yields mAs

\[ I_{\text{tube}} = 1-1000\text{mA} \]

mAs measures x-ray exposure
X-Ray Spectra

- Bremmstrahlung (x-rays within anode)
- Leaving anode
- Leaving tube
- Characteristic radiation
- After filter
- Leaving body

Relative intensity vs. photon energy, keV
Bremsstrahlung

- Continuous spectrum of EM radiation is produced by abrupt deceleration of charged particles ("Bremsstrahlung" is German for "braking radiation").

- Deceleration is caused by deflection of electrons in the Coulomb field of the nuclei.

- Most of the energy is converted into heat, ~0.5% is x-ray.

- The energy of the generated x-ray photon is given by energy conservation:

\[ h\nu = K_e - K'_e \]

- The maximum energy for the produced photon is given by:

\[ E_{p,\text{max}} = h\nu = K_e = eV_{\text{tube}} \]

[From Graber, Lecture Note for BMI1-FS05]
Bremsstrahlung intensity

- Overall Bremsstrahlung intensity $I$:

\[ I \propto V_{\text{tube}}^2 I_{\text{tube}} \]

Electrical power consumption of tube: $P_{\text{tube}} = I_{\text{tube}} \times V_{\text{tube}}$ [W]

- The produced x-ray power $P_x$ (in[W]) is given by:

\[ P_x = k Z V_{\text{tube}}^2 I_{\text{tube}} = kZ V_{\text{tube}} P_{\text{tube}} = \eta P_{\text{tube}} \]

\[ \eta = \frac{P_x}{P_{\text{tube}}} = kZ V_{\text{tube}} \]  

- x-ray production efficiency

- Material constant $k = 1.1 \times 10^{-9}$ for Tungsten (Z=74).

[From Graber, Lecture Note for BMI1-FS05]
Bremsstrahlung spectrum

- Theoretically, bremsstrahlung from a thick target creates a continuous spectrum from $E = 0$ to $E_{\text{max}}$ with intensity $I_b$:

$$I_b(E) \sim Z(E_{\text{max}} - E)$$

- Actual spectrum deviates from ideal form due to:
  - Absorption in window / gas envelope material and absorption in anode
  - Multienergetic electron beam

[From Graber, Lecture Note for BMI1-FS05]
Characteristic radiation

- Narrow lines of intense x-ray at characteristic energies are superimposed on the continuous bremsstrahlung spectrum.
- Caused by removal of inner shell electrons and subsequent filling of hole with electrons from higher shell. The shell-energy difference determines the energy of characteristic rays.
- Lines are named after the lower shell involved in the process; the upper shell involved is denoted by Greek letters:
  \[ \Delta n = 1 \rightarrow \alpha\text{-transitions}, \Delta n = 2 \rightarrow \beta\text{-transitions}, \ldots \]

[From Graber, Lecture Note for BMI1-FS05]
Different types of characteristics rays

From http://hyperphysics.phy-astr.gsu.edu/Hbase/quantum/xterm.html#c1
X-ray spectra

- X-ray for general diagnostic radiology produced at 40 – 150 kVp
- Maximum photon energy: 
  \[ E_p [\text{keV}] = h \nu_{\text{max}} = e \times \text{kVp} \]
- Characteristic radiation occurs only for anode voltages 
  \[ e \times \text{kVp} > I_{K,L,M,...} \]

[From Graber, Lecture Note for BMI1-FS05]
X-ray tube design

- Cathode w/ focusing cup, 2 filaments (different spot sizes)
- Anode
  - Tungsten, $Z_w = 74$, $T_{\text{melt}} = 2250$ °C
  - Embedded in copper for heat dissipation
  - Angled (see next slide)
  - Rotating to divert heat

[From Graber, Lecture Note for BMI1-FS05]
Filtration

- Low energy x-ray will be absorbed by the body, without providing diagnostic information
- Filtration: Process of absorbing low-energy x-ray photons before they enter the patient
  - Inherent filtration
    - Within anode
    - Glass housing
  - Added filtration
    - Aluminum
    - Copper/Aluminum
      Note: Cu has 8keV characteristic x-rays
    - Measured in mm Al/Eq
**Goal:** To direct beam toward desired anatomy
Compensation Filters

- Goal: to even out film exposure
Contrast Agents

Goal: To create contrast where otherwise none

When the x-ray energy exceeds the Kedge (binding energy of K-shell), the mu coefficient is much higher, providing high contrast.
• **Iodine:**
  - Can be synthesized into soluble compounds that are safely introduced through intravascular injection or ingestion
  - Used for imaging of
    • Blood vessels, heart chambers, tumors, infections
    • Kidneys, bladder
  - Naturally exist in thyroid, and hence X-ray is very good for thyroid imaging

• **Barium**
  - Administered as a “chalky milkshake”
  - Used in the gastrointestinal tract,
    • Stomach, bowel

• **Air**
  - Does not absorb x-ray
  - “opposite” type of contrast
  - By Inflating the lungs, air provides contrast for lung tissues
Scatter Control

- Ideal x-ray path: a line!
- Compton scattering causes blurring
- How to reduce scatter?
  - airgap
  - scanning slit
  - grid
Grids

- Effectiveness in scatter reduction?

\[ \text{grid ratio} = \frac{h}{b} \]

- 6:1 to 16:1 (radiography) or 2:1 (mammo)
Problem with Grids

- Radiation is absorbed by grid
  - grid conversion factor
    \[
    GCF = \frac{\text{mAs w/ grid}}{\text{mAs w/o grid}}
    \]
  - Typical range $3 < GCF < 8$
- Grid visible on x-ray film
  - move grid during exposure
  - linear or circular motion
Film-Screen Detector

- Film stops only 1–2% of x-rays
- Film stops light really well

Intensifying screen:

Phosphor: convert x-ray photons to light

Reflective layer: Reflect light back to film

- Phosphor = calcium tungstate
- Flash of light lasts $1 \times 10^{-10}$ second
- $\sim 1,000$ light photons per 50 keV x-ray photon
Radiographic Cassette
Digital Radiology

- Replace the intensifying screen/X-ray film by
  - flat panel detectors (FPD) using thin-film transistor (TFT) arrays
  - A scintillator
    - Consisting of many thin, rod-shaped cesium iodide (CsI) crystals
- When an X-ray is absorbed in a CsI rod, the CsI scintillates and produces light
- The light is converted into an electrical signal by a photodiode in the TFT array
- The electrical signal is amplified and converted to a digital value using an A/D converter
- A typical commercial DR system has flat panel dimensions of 41x41 cm, with an TFT array of 2048x2048 elements
- Ref: Webb, Introduction to biomedical imaging, Sec. 1.5.5
Biological effects of ionizing radiation

- Damage depends on deposited (= absorbed) energy (intensity × time) per tissue volume
- Threshold: No minimum level is known, above which damage occurs
- Exposure time: Because of recovery, a given dose is less harmful if divided
- Exposed area: The larger the exposed area the greater the damage (collimators, shields!)
- Variation in Species / Individuals: LD 50/30 (lethal for 50% of a population over 30 days, humans ~450 rads / whole body irradiation)
- Variation in cell sensitivity: Most sensitive are nonspecialized, rapidly dividing cells (Most sensitive: White blood cells, red blood cells, epithelial cells. Less sensitive: Muscle, nerve cells)
- Short/long term effects: Short term effects for unusually large (> 100 rad) doses (nausea, vomiting, fever, shock, death); long term effects (carcinogenic/genetic effects) even for diagnostic levels ⇒ maximum allowable dose 5 R/yr and 0.2 R/working day [Nat. Counc. on Rad. Prot. and Meas.]

[From Graber, Lecture Note for BMI1-FS05]
Image Formation

• Basic imaging equation
• Geometric effects
• Extended source
• Film blurring
• Impact of noise and scattering
Basic Imaging Equations

\[ I(x, y) = \int_0^\infty S_0(E')E' \exp \left\{ - \int_0^{r(x,y)} \mu(s; E', x, y)ds \right\} dE' \]
1) What is the local contrast of the blood vessel?
2) What is the local contrast of the blood vessel when contrast agent is injected?

Blood vessel $\mu=0.2$
w/contrast $\mu=20$

Soft tissue $\mu=0.4$

w/o contrast:

$\begin{align*}
I_b &= I_{\min} = I_0 e^{-(0.4*2.0)}; \\
I_o &= I_{\max} = I_0 e^{-(0.4*1.5+0.2*0.5)}
\end{align*}$

Local contrast: $C_1 = \frac{I_o-I_b}{I_b}$

Global contrast: $C = \frac{I_{\max}-I_{\min}}{I_{\max}+I_{\min}}$

w/ contrast:

$\begin{align*}
I_b &= I_{\max} = I_0 e^{-(0.4*2.0)}; \\
I_o &= I_{\min} = I_0 e^{-(0.4*1.5+20*0.5)}
\end{align*}$

Local contrast: $C_1 = \frac{|I_o-I_b|}{I_b}$

Global contrast: $C = \frac{I_{\max}-I_{\min}}{I_{\max}+I_{\min}}$
Geometric Effects

- X-rays are diverging from source
- Undesirable effects:
  - $\cos^3 \theta$ falloff across detector
  - anode heel effect
  - pathlength irregularities
  - magnification
- $I_0$ is intensity at $(0, 0)$
- $r$ is distance from $(x, y)$ to x-ray origin
- $\theta$ is angle between $(0, 0)$ and $(x, y)$
Inverse Square Law

- Net flux of photons decrease as $1/r^2$.
  
  Therefore
  
  $$I_0 = \frac{I_S}{4\pi d^2} \quad I_r = \frac{I_S}{4\pi r^2}$$

- Eliminate source intensity $I_S$
  
  $$I_r = I_0 \frac{d^2}{r^2}$$

- Since $\cos \theta = d/r$
  
  $$I_r = I_0 \cos^2 \theta$$

$I_0$ is the detected flux at the origin of the detector plane
$I_r$ is the detected flux at an arbitrary point of the detector plane with angle $\theta$ w/o considering the oblique effect discussed in the next page.
Obliquity

\[ I_d = I_0 \cos \theta \]

- Intensity is

\( I_0 \) should be replaced by \( I_r \)
Overall Effect of Beam Divergence

- Inverse square law and obliquity combine

\[ I_d(x_d, y_d) = I_0 \cos^3 \theta \]

- Can usually be ignored. Why?
  - Detector is far away
  - Field of view (FOV) is often small

\[ \Rightarrow \theta \text{ is small} \]
Anode Heel Effect

- Intensity within the x-ray cone
  - Not uniform
  - stronger in the cathode direction
  - 45% variation is typical
- Compensate, use to advantage, or ignore
- We will ignore in math
Imaging of a Uniform Slab

- Uniform slab yields different intensities
• Intensity on detector

\[ I_d(x, y) = I_0 \exp\left\{-\mu L / \cos \theta\right\} \]

• Including inverse square law and obliquity:

\[ I_d(x, y) = I_i \cos^3 \theta \exp\left\{-\mu L / \cos \theta\right\} \]

• If \( d \approx r \) all effects can be ignored

\[ I_i = I_s / (4 \pi d^2) \]

Illustrate the received intensity as function of \( y \) or \( x \) or \( \theta \)
Received Signal as a Function of Theta

\[ \cos(q)^3 \exp(-1/\cos(q)) \]

This plot assumes \( \mu L = 1 \), e.g. \( \mu = 1/\text{cm}, L = 1\text{cm} \).
• How does it vary as a function of Y when x=0 (vertical axis of the detector plain)?
• \( \cos(q) = \frac{d}{r} = \frac{d}{\sqrt{d^2 + y^2}} / \)
• Assuming \( d=5 \text{m}, y= -10 \text{cm to 10cm} \) (\( q \) from 0 to 1.14 degree)
• Vary small relative change in the range of \( y \)

Intensity as function of \( y \)
Example: Image of a prism due to a point source

Consider the x-ray imaging of a cube. Determine the intensity of detected photons along the y axis on the detector plane. Express your solution in terms of the angle $q$. Sketch this function. You should consider the inverse square law and the oblique effect. Assume the x-ray source is an ideal point source with intensity $I_0$, and the object has a constant linear attenuation coefficient $m$. (Example 5.4 in textbook)
Solution

Sketch over in class. Also see textbook
Must consider different regions separately
Objects Magnification

- Size on detector depends on distance from source

\[ w_z = w \frac{d}{z} \]

Magnification factor:

\[ M(z) = \frac{d}{z} \]
Imaging of a Thin Non-Uniform Slab

- Assume a very thin slab at z
  - the linear absorption coefficient at \((x',y')\) is \(\mu(x',y')\)
  - Detector position \((x,y)\) -> slab position \((x',y')\)

\[
x/d = x'/z \rightarrow x' = x \frac{z}{d} = x / M(z)
\]
- Let “transmittivity” be

\[ t_z(x, y) = \exp\{-\mu(x, y)\Delta z\} \]

- On detector, intensity is

\[ I_d(x, y) = I_0 \cos^3 \theta t_z \left( \frac{x}{M(z)}, \frac{y}{M(z)} \right) \]

- After substitution

\[ I_d(x, y) = I_0 \left( \frac{d}{\sqrt{d^2 + x^2 + y^2}} \right)^3 t_z \left( \frac{xz}{d}, \frac{yz}{d} \right) \]

\[ I_0 = I_s / (4 \pi d^2) \]
Blurring Due to Extended Source

First study the image through a pinhole
- Impulse response

Image through an arbitrary objects
- Impulse response * object attenuation profile
Image of source through a pinhole

Pinhole: a infinitesimal hole (area=0) that passes the X-ray source w/o attenuation. Everywhere else the X-ray is completely absorbed.

Reversed image

\[ D' / (d-z) = -D / z \rightarrow D' = -D (d-z)/z = Dm \]

- Source magnification:

\[ m(z) = \frac{d - z}{z} = 1 - M(z) \]

Call the following \( h(x,y) \) (response due to a pinhole at (0,0)

- Image of source through pinhole at \( z \)

\[ I_d(x, y) = \frac{\cos^3 \theta}{4\pi d^2 m^2} s \left( \frac{x}{m}, \frac{y}{m} \right) \]

Loss of source intensity due to inverse square law

Scale factor due to pinhole (See textbook)
Image of an Arbitrary Slice

- An arbitrary slab at \( z \) can be thought of as many pinholes at different locations \((x', y')\), each with transmittivity \( t_z(x', y') \)
  - The received signal due to transmittivity at \((x', y')\) can be written as \( h(x-x', y-y') \) \( t_z(x', y') \) assuming the system is translation invariant

- The image of the slab is a sum of individual images of the source through all the pinholes multiplied by the respective transmittivity
  - \( I_d(x, y) = \int_{x', y'} h(x-x', y-y') t_z(x', y') \, dx' \, dy' \)

- The overall effect can be captured through linear convolution

\[
I_d(x, y) = \frac{\cos^3 \theta}{4\pi d^2 m^2} t_z \left( \frac{x}{M}, \frac{y}{M} \right) \ast s \left( \frac{x}{m}, \frac{y}{m} \right)
\]

Note: \( m \) depends on \( z \), distance of slab to the source
Example

- Source is a circular disk with diameter D
- Object is square plane with dimension W at distance z
- Detector plane at distance d from source
- How does the detected image look for d=2Z and d=3Z
- Note that the blurring of the edge depends on z

- What is \( t_z(x,y) \) and \( s(x,y) \)?
- What is \( I_d(x,y) \)?
- How is \( I_d(x,y) \) related with \( t_z(x,y) \)?
- How does the image of \( I_d(x,y) \) look?
Example: solution

$T_z(x,y)$: a square with width $W$
$S(x,y)$: a disk with diameter $D$. Assuming $D << W$,

- **For** $d=2z$, $M=d/z=2$, $m=1-M=-1$
  
  $T_z(x/M,y/M)$: a square with width $2W$
  $S(x/m,y/m)$: a disk with diameter $D$
  
  The resulting detector image is a square with width $2W$ but with a blurred edge with blurring width $D$

- **For** $d=3z$, $M=d/Z=3$, $m=1-M=-2$
  
  $T_z(x/M,y/M)$: a square with width $3W$
  $S(x/m,y/m)$: a disk with diameter $2D$
  
  The resulting detector image is a square with width $3W$ but with a blurred edge with blurring width $2D$
For the previous example, L is very small, but the source has diameter D, blurring is due to the source diameter being non-zero.
A single x-ray photon causes a blurry spot on the film which is effectively the “impulse response” to the x-ray impulse $h(x,y)$

Typical MTF for a film-screen detector
Overall Imaging Equation

- Including all effects (geometric, extended source, film-screen blurring), the image corresponding to a slab at \( z \) with transmittivity function \( t_z(x,y) \) is

\[
I_d(x, y) = \cos^3 \theta \frac{1}{4\pi d^2 m^2} s \left( \frac{x}{m}, \frac{y}{m} \right) \star t_z \left( \frac{x}{M}, \frac{y}{M} \right) \star h(x, y)
\]

- For an object with a certain thickness, the transmittivity function must be modified to reflect the overall attenuation along the \( z \)-axis.
- When the source is polyenergetic, integration over photon energy is additionally needed.
Example

• In the previous example, how would the image look if the film blurring is a box function of width $h$?
Film Characteristics

- Film darkening (after development) depends on incident light (which depends on the incident x-ray)
- Optical density

\[ D = \log_{10} \frac{I_i}{I_t} \]

- Usable densities \(0.25 < D < 2.25\)
- Best densities \(1.0 < D < 1.5\)
Optical Density vs. Exposure

- X-ray exposure yields optical density
  \[ D = \Gamma \log_{10} \frac{X}{X_0} \]
- \( \Gamma \) is film gamma
- Typical ranges: \( 0.5 < \Gamma < 3.0 \)
- Latitude is range exposures where relationship is linear
- Speed is inverse of exposure at which
  \[ D = 1 + \text{fog level} \]
The H&D Curve

- (a) High speed film with CaWO₄ screens
- (b) Direct x-ray film
- (c) High-speed film without screens

Optical density vs. Exposure, mR
Effect of Noise

• Source of noise:
  – Detector does not faithfully reproduce the incident intensity
  – X-rays arrive in discrete packets of energy. This discrete nature can lead to fluctuations in the image

• Local contrast
  \[ C = \frac{I_t - I_b}{I_b} \]

• Signal is \( I_t - I_b \)

• Noise is due to Poisson behavior

• Variance of noise in background: \( \sigma_b^2 \)

• Signal to noise
  \[ \text{SNR} = \frac{I_t - I_b}{\sigma_b} = \frac{C I_b}{\sigma_b} \]
How is noise related to signal?

- Assuming the number of photons in each burst follows the Poisson distribution
  - \( P(N=k) = \frac{(a^k)}{k!} \cdot e^{-a} \)
  - Variance = mean = \( a \)

- Let \( N_b \) denotes the average number of photons per burst per area
- Let \( h\nu \) denotes the effective energy for the X-ray source
- The average background intensity is

\[
I_b = \frac{N_b h\nu}{A\Delta t}
\]

- The variance of photon intensity is

\[
\sigma^2_b = N_b \left( \frac{h\nu}{A\Delta t} \right)^2
\]

- The SNR is

\[
\text{SNR} = C \sqrt{N_b}
\]

- SNR can be improved by
  - Increasing incident photon count
  - Improving contrast

\[
Y = aX \text{ is a RV with mean } \eta_y = a\eta_x, \text{ variance } \sigma^2_y = a^2\sigma^2_x
\]
Detective Quantum Efficiency

- How good is a detector?
- Consider:
  - Potential SNR before detection
  - Actual SNR upon detection
- Detective Quantum Efficiency
  \[
  DQE = \left( \frac{SNR_{\text{out}}}{SNR_{\text{in}}} \right)^2
  \]
- Degradation of SNR during detection

When a x-ray source has mean intensity $m = N_b$, and variance $s^2 = N_b$, $SNR = m/s = \sqrt{N_b}$
Example

• Suppose an X-ray tube is set up to fire bursts of photons each with N=10000 photons and the detector’s output (# of detected photons per burst) x has a mean =8000, variance=40000. What is its DQE?

• Solution:

The actual # of photons fired at the x-ray tube follows the Poisson process (mean = variance = 10000)

\[ \text{SNR}_{\text{in}} = \frac{\text{mean}}{\sqrt{\text{variance}}} = \sqrt{10000} = 100 \]

The # of detected photons has mean = 8000, variance = 40000

\[ \text{SNR}_{\text{out}} = \frac{\text{mean}}{\sqrt{\text{variance}}} = \frac{8000}{\sqrt{40000}} = \frac{8000}{200} = 40 \]

\[ \text{DQE} = \left( \frac{\text{SNR}_{\text{out}}}{\text{SNR}_{\text{in}}} \right)^2 = 0.16 \]

This means that only about 16% of photons are detected correctly
Effect of Compton Scattering

• Compton scattering causes the incident photons to be deflected from their straight line path
  – Add a constant intensity $I_s$ in both target and background intensity ("fog")
  – Decrease in image contrast
  – Decrease in SNR

W/o scattering :

<table>
<thead>
<tr>
<th>term</th>
<th>expression</th>
<th>unit</th>
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<tbody>
<tr>
<td>target intensity $I_t$</td>
<td>$I_t$</td>
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<tr>
<td>background intensity $I_b$</td>
<td>$I_b$</td>
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<tr>
<td>contrast $C$</td>
<td>$\frac{I_t - I_b}{I_b}$</td>
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<td>SNR $= C \frac{I_b}{\sigma_b}$</td>
<td>$= C \sqrt{N_b}$</td>
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<td>$I_t + I_s$</td>
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<td>background intensity $I_b$</td>
<td>$I_b + I_s$</td>
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<tr>
<td>contrast $C'$</td>
<td>$\frac{I_t - I_b}{I_b + I_s}$</td>
<td>$= \frac{I_b}{I_b + I_s} \frac{I_t - I_b}{I_b}$</td>
</tr>
<tr>
<td>SNR $' = C \frac{I_b}{\sigma_b}$</td>
<td>$= C \frac{N_b}{\sqrt{N_b + N_s}} = C \frac{\sqrt{N_b}}{\sqrt{1 + N_s / N_b}} = \text{SNR} \frac{1}{\sqrt{1 + I_s / I_b}}$</td>
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\[ \text{SNR\'} = C \frac{I_b}{\sigma_b} = C \frac{N_b}{\sqrt{N_b + N_s}} = C \frac{\sqrt{N_b}}{\sqrt{1 + N_s / N_b}} = \text{SNR} \frac{1}{\sqrt{1 + I_s / I_b}} \]
Medical Applications

- Orthopedic
- Chest
- Abdomen
- Mammography
- Angiography

[From Graber, Lecture Note for BMI1-FS05]
Mammography

- Detection and diagnosis (symptomatic and screening) of breast cancer
- Pre-surgical localization of suspicious areas
- Guidance of needle biopsies.

- Breast cancer is detected on the basis of four types of signs on the mammogram:
  - Characteristic morphology of a tumor mass
  - Presentation of mineral deposits called microcalcifications
  - Architectural distortions of normal tissue patterns
  - Asymmetry between corresponding regions of images on the left and right breast

⇒ Need for good image contrast of various tissue types.
- Simple x-ray shadowgram from a quasi-point source.

[From Graber, Lecture Note for BMI1-FS05]
Mammography contrast

- Image contrast is due to varying linear attenuation coefficient of different types of tissue in the breast (adipose tissue (fat), fibroglanular, tumor).
- Ideal energy distribution of X-ray should be below 20 for average size breast, slightly higher for denser breast.

[From Graber, Lecture Note for BMI1-FS05]
Mammography source

- Voltage ~ 25-30 kVp
- Anode material Mo (Molybdenum), Rh (Rhodium) (characteristic peaks at 17.9 and 19 for Mo, and slightly higher for Rh)
- Filtering: use Mo or Rh to absorb energy above 20 or 25Kev

[From Graber, Lecture Note for BMI1-FS05]
Anti-scatter grid

- Significant Compton interaction for low $E_p$ (37-50% of all photons).
- Linear grid: Lead septa + interspace material. Septa focused toward source. Grid ratio $\sim 3.5$-5:1. Only scatter correction in one dimension. Scatter-to-primary (SPR) reduction factor $\sim 5$
- Recently crossed grid introduced
- Grids are moved during exposure
- Longer exposure

[From Graber, Lecture Note for BMI1-FS05]
X-ray projection angiography

- Imaging the circulatory system. Contrast agent: Iodine (Z=53) compound; maximum iodine concentration ~ 350 mg/cm³
- Monitoring of therapeutic manipulations (angioplasty, atherectomy, intraluminal stents, catheter placement).
- Short intense x-ray pulses to produce clear images of moving vessels. Pulse duration: 5-10 ms for cardiac studies ...100-200 ms for cerebral studies
Summary

• Projection radiography system consists of an x-ray tube, devices for beam filtration and restriction, compensation filters, grids, and a film-screen detector (or digital detector, filmless)
• The detector reading (or image gray level) is proportional to the number of unabsorbed x-ray photons arriving at the detector, which depends on the overall attenuation in the path from the source to the detector
• The above relation must be modified to take into account of inverse square law, obliquity, anode heel effect, extended source and detector impulse response
• The degree of film darkening is nonlinearly related to the film exposure (detected x-ray) by the H&D curve
• Both detector noise and Compton scattering reduce contrast and SNR of the formed image
Reference

- Prince and Links, Medical Imaging Signals and Systems, Chap 5.
- Webb, Introduction to biomedical imaging, Chap 1.
Homework

• Reading:
  – Prince and Links, Medical Imaging Signals and Systems, Chap 5.
• Note down all the corrections for Ch. 5 on your copy of the textbook based on the provided errata.
• Problems for Chap 5 of the text book:
  – P5.2
  – P5.4
  – P5.5
  – P5.8
  – P5.18
  – P5.19
  – correction: the sentence “Suppose a 5 cm …” in Part (a) should be moved to the beginning of part (b). Also, intrinsic contrast in part (b)= \((\mu_t-\mu_b)/(\mu_t+\mu_b)\), contrast in part (c)= \((I_{\text{max}}-I_{\text{min}})/(I_{\text{max}}+I_{\text{min}})\).
  – P5.22
1. Consider the x-ray imaging of a two-layer slab, illustrated below. Determine the intensity of detected photons along the y axis on the detector plane. Express your solution in terms of the y-coordinate. Sketch this function. You should consider the inverse square law and the oblique effect. Assume the x-ray source is an ideal point source with intensity $I_0$. For simplicity, assume the slab is infinitely long in the y direction.