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Medical Imaging

Projection Radiography

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Based on Prince and Links, Medical Imaging Signals and Systems and Lecture Notes by Prince. Figures are from the book.
Lecture Outline

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

- x-ray tube and housing
- collimator
- x-rays
- patient
- table
- ionization chamber
- grid
- cassette
- Bucky assembly
X-ray Tube
Another diagram of X-ray tube
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

SCR: Silicon-controlled rectifier, switch timed by a processor
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]
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.
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: \( E_p[\text{keV}] = \text{Tube Voltage (eV)} \)

- Characteristic radiation occurs only for anode voltages > binding energy of electrons in inner shells

[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 (metal placed outside the tube)
  – Aluminum (standard)
  – Copper (for higher energy systems)
  – Copper must be followed by aluminum to absorb the 8KeV characteristic x-rays created within the copper
Beam Hardening

Bremmstrahlung
(x-rays within anode)

leaving anode

leaving tube

characteristic radiation

after filter

leaving body

relative intensity

photon energy, keV
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 (33.2 KeV)

Barium (37.4 KeV)
Contrast Agents in Clinical Use

- **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
• Effectiveness in scatter reduction?

$$\text{grid ratio} = \frac{h}{b}$$

• 6:1 to 16:1 (radiography) or 2:1 (mammo)

• What is the maximum scattering angle given $h$ and $b$?
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 Radiography

- Replace film by digital detectors
- Earlier systems: Computer Radiography (CR)
  - Store latent images in photostimulable imaging plates (PSP)
  - Need to be scanned by laser to form a digital image.
- CCD-based system
  - Small field of view
  - Group multiple CCD arrays
- Thin-film-transistor (TFT) based system
- CMOS-based system
  - Low cost: widely used in consumer cameras
  - Small field of view
  - Easier to tie multiple substrates than CCD arrays
- Portable, and reusable, some have wireless transmitters built in to send captured image to servers
TFT-Based 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 X-ray photons are 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 x 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

\[ I(x, y) = \int_0^\infty S_0(E')E' \exp \left\{ - \int_0^{r(x,y)} \mu(s; E', x, y) \, ds \right\} \, dE' \]
Example

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?

\[
\begin{align*}
\text{Blood vessel } \mu &= 0.2 \\
\text{w/o contrast:} \\
I_b &= I_{\text{min}} = I_0 e^{-(0.4 \times 2.0)}; \\
I_o &= I_{\text{max}} = I_0 e^{-(0.4 \times 1.5 + 0.2 \times 0.5)} \\
\text{Local contrast: } C_1 &= \frac{I_o - I_b}{I_b}; \\
\text{Global contrast: } C &= \frac{I_{\text{max}} - I_{\text{min}}}{I_{\text{max}} + I_{\text{min}}} \\
\text{w/ contrast:} \\
I_b &= I_{\text{max}} = I_0 e^{-(0.4 \times 2.0)}; \\
I_o &= I_{\text{min}} = I_0 e^{-(0.4 \times 1.5 + 20 \times 0.5)} \\
\text{Local contrast: } C_1 &= \frac{|I_o - I_b|}{I_b}; \\
\text{Global contrast: } C &= \frac{I_{\text{max}} - I_{\text{min}}}{I_{\text{max}} + I_{\text{min}}}
\end{align*}
\]
**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 \((x, y)\)
Variation Due to 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

\[ I_d = I_0 \cos \theta \]

\( I_0 \) 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
Anode Heel Effect

From http://79.170.44.110/hullrad.org.uk/openppt/FRCR/Diagnostic Radiology - Tim Wood/Lecture 2 - X-ray tube.ppt
Imaging of a Uniform Slab

- Uniform slab yields different intensities
- Intensity on detector

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

- Including inverse square law and obliquity:

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

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

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

\[ \cos(\theta)^3 \exp\left(-\frac{1}{\cos(\theta)}\right) \]

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)?
• \( \cos(\theta) = \frac{d}{r} = \frac{d}{\sqrt{d^2 + y^2}} \)
• Assuming \( d=5m, y= -10cm \) to 10cm (\( \theta \) from 0 to 1.14 degree)
• Very small relative change in the range of y

intensity as function of y
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.
Solution

• Sketch on the board
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 $\theta$. 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 $\mu$. (Example 5.4 in textbook)
Solution

Sketch over in class. Also see textbook.
Complete as homework
Must consider different regions separately
Note the edge blurring effect.
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 \((x', y')\)

\[
x / d = x' / z \rightarrow x' = x \frac{z}{d} = x / M(z)
\]
\[
x = x' M(z)
\]
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

Note: blurring depends on the object distance! and source size. Less blurring when the object is closer to the detector plane!
Image of source through a pinhole

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

Reversed image

\[ \frac{D'}{(d-z)} = -\frac{D}{z} \rightarrow D' = -D \frac{(d-z)}{z} = Dm \]

- 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 \(z\)

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

Scale factor due to pinhole (See textbook)

Loss of source intensity due to inverse square law
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 $t_z(x', y') = \exp(-m(x', y')L)$
  - 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
- The overall effect can be captured through linear convolution

$$I_d(x, y) = \int_{x', y'} h(x-x', y-y') t_z(x', y') dx' dy'$$

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

Note: $m$ and $M$ depends on $z$, distance of slab to the source. When the object is next to the detector $M=1$, less blurring.
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?
Solution

- By sketch
- By using the convolution formula
Example: solution

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

- For \( d=2z \), \( M=d/z=2 \), \( m=1-M=-1 \)
  \( Tz(x/M,y/M) \): 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 \( d=3z \), \( M=d/Z=3 \), \( m=1-M=-2 \)
  \( Tz(x/M,y/M) \): 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 \( 2D \)
Another Example

- Prob. 5.23
- Source
  \[ s(x, y) = s_0 e^{-x^2} \delta(y) \]
- Object to be imaged is a line phantom, with a transmission function given by
  \[ t(x, y) = \delta(x - \frac{w}{2}) + \delta(x + \frac{w}{2}) \]
- How would the detected image look?
- Determine the minimal value of \( w \) such that the images of the two lines on the detector plane can be distinguished
Example
Digital Detector Blurring

- Digital detectors lead to blurring in the detected image, due to the thickness of the scintillators.
- Such effect can be modeled by another impulse response, denoted by $h(x,y)$. 
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

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

• 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
What does it mean?

• Output image = scaled version of the object * g(x,y)

• Scaling depends on the object position z
  – Less magnification when object is closer to the detector

• g(x,y)= ?

• g(x,y) depends on both source and detector blurring and position z!
Example

• In the previous example, how would the image look if the film blurring is a box function of width h?
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**
  \[ \text{SNR} = \frac{I_t - I_b}{\sigma_b} = \frac{CI_b}{\sigma_b} \]
How is noise related to signal?

- Assuming the number of photons in each burst follows the Poisson distribution
  - \( P(N=k) = \left(\frac{a^k}{k!}\right) e^{-a} \)
  - 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
  \[
  \sigma_b^2 = N_b \left(\frac{h \nu}{A \Delta t}\right)^2
  \]

- The SNR is
  \[
  \text{SNR} = C \sqrt{N_b}
  \]

- SNR can be improved by
  - Increasing incident photon count
  - Improving contrast
- Recall SNR = \( \frac{I_t - I_b}{s_b} = C \frac{I_b}{s_b} \), C = \( \frac{I_t - I_b}{I_b} \)
Detective Quantum Efficiency

- How good is a detector?
- Consider:
  - Potential SNR before detection
  - Actual SNR upon detection
- Detective Quantum Efficiency

\[
DQE = \left( \frac{SNR_{out}}{SNR_{in}} \right)^2
\]

- 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)

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

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

Without scattering:
- Target intensity: $I_t$
- Background intensity: $I_b$
- Contrast: $C = \frac{I_t - I_b}{I_b}$
- SNR: $\frac{I_b}{\sigma_b} = C \sqrt{N_b}$

With scattering:
- Target intensity: $I_t + I_s$
- Background intensity: $I_b + I_s$
- Contrast: $C' = \frac{I_t - I_b}{I_b + I_s} = \frac{I_b}{I_b + I_s} C = \frac{C}{1 + \frac{I_s}{I_b}}$
- SNR': $\frac{I_b}{\sigma_b} = C \frac{N_b}{\sqrt{N_b + N_s}} = C \frac{\sqrt{N_b}}{\sqrt{1 + \frac{N_s}{N_b}}} = \text{SNR} \frac{1}{\sqrt{1 + \frac{I_s}{I_b}}}$
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), fibro glandular, tumor).
- Ideal energy distribution of X-ray should be below 20KeV 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 25 Kev

[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 ~ 3.5-5:1. Only scatter correction in one dimension. Scatter-to-primary (SPR) reduction factor ~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

[From Graber, Lecture Note for BMI1-FS05]
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 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.
Imaging Equations

• Basic imaging equation (relating detector reading with attenuation coefficient)

\[ S(x; E) = S_0(E) \exp \left\{ - \int_0^x \mu(x'; E) \, dx' \right\} \]

• Modified to consider inverse square law, obliquity, anode heel effect

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

• Further modified to consider extended source and detector impulse response

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

\[ \ast \ t_z \left( \frac{x}{M}, \frac{y}{M} \right) \ast h(x, y) \]
Reference

- Prince and Links, Medical Imaging Signals and Systems, Chap 5.
- Webb, Introduction to biomedical imaging, Chap 1.
Homework Lecture 3

- Reading:
  - Prince and Links, Medical Imaging Signals and Systems, Chap 5.
- Problems for Chap 5 of the text book:
  - P5.2
  - P5.4
  - P5.5
  - P5.9
  - P5.18
  - P5.22
  - P5.23 (You could use the general imaging equation involving convolution. But you could also derive the solution just by geometric sketching!)
  - P5.25