EL5823/BE6203 Medical Imaging

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.

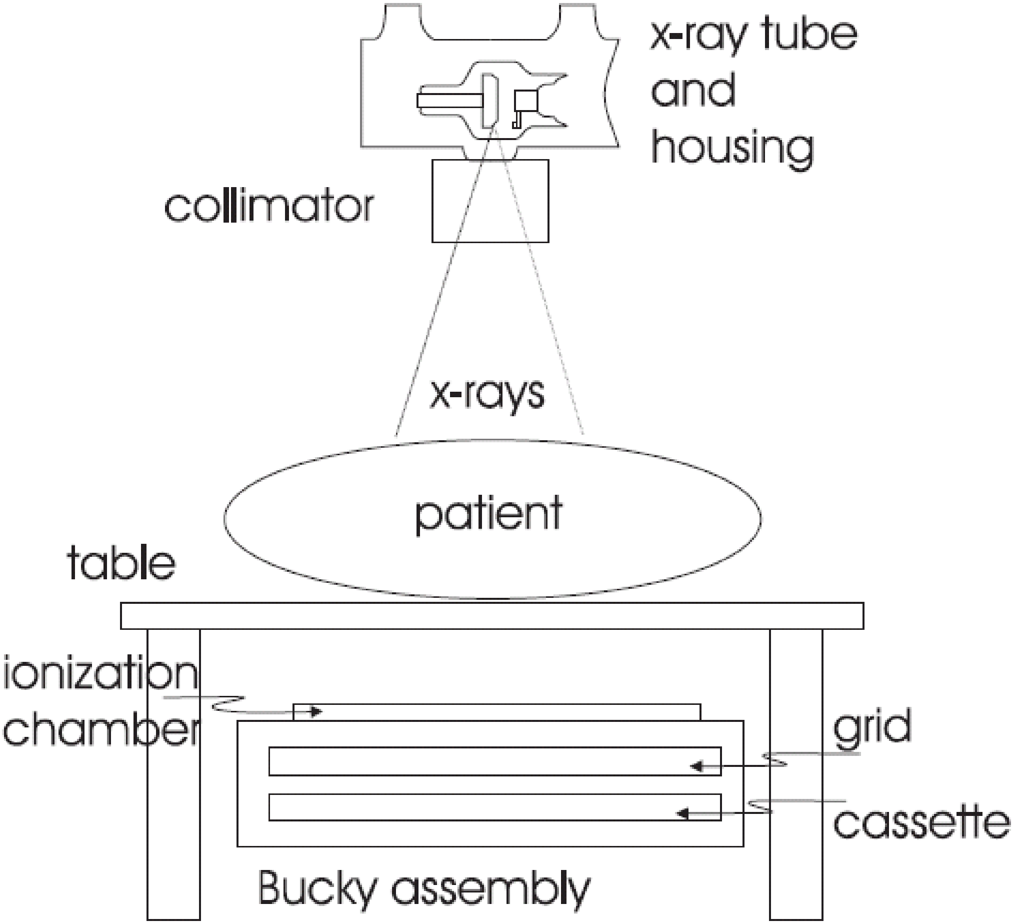
Lecture Outline

- 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

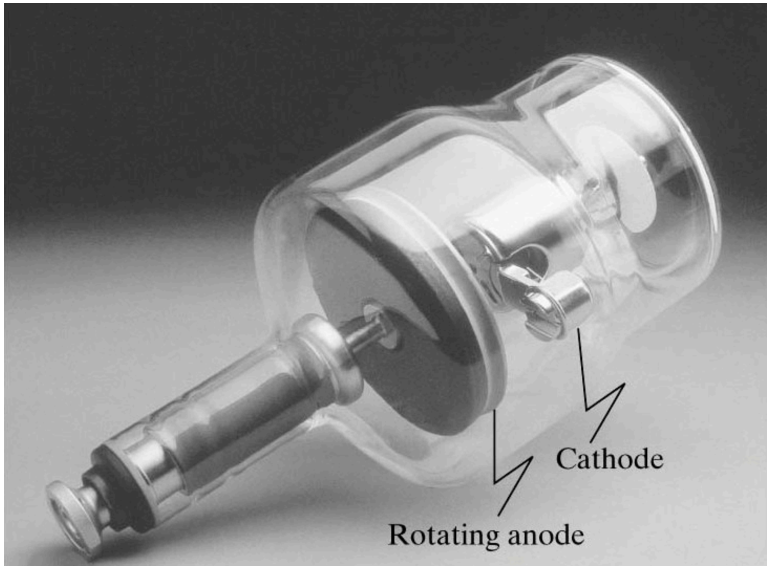
Overview

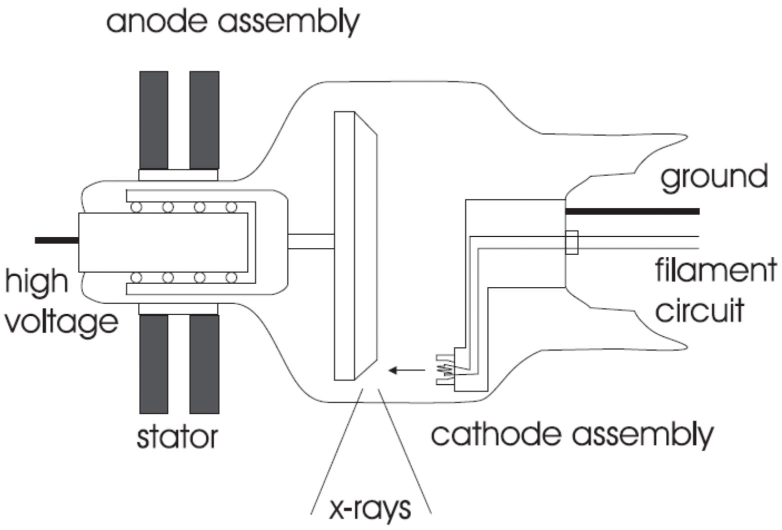
- Systems:
  - chest x-rays, mammography
  - dental x-rays
  - fluoroscopy, angiography
- Properties
  - high resolution
  - low dose
  - broad coverage
  - short exposure time

Radiographic System



X-ray Tube


Cathode
Rotating anode


anode assembly
ground
high
voltage
filament
circuit
stator
cathode assembly
x-rays

X-Ray Tube Components

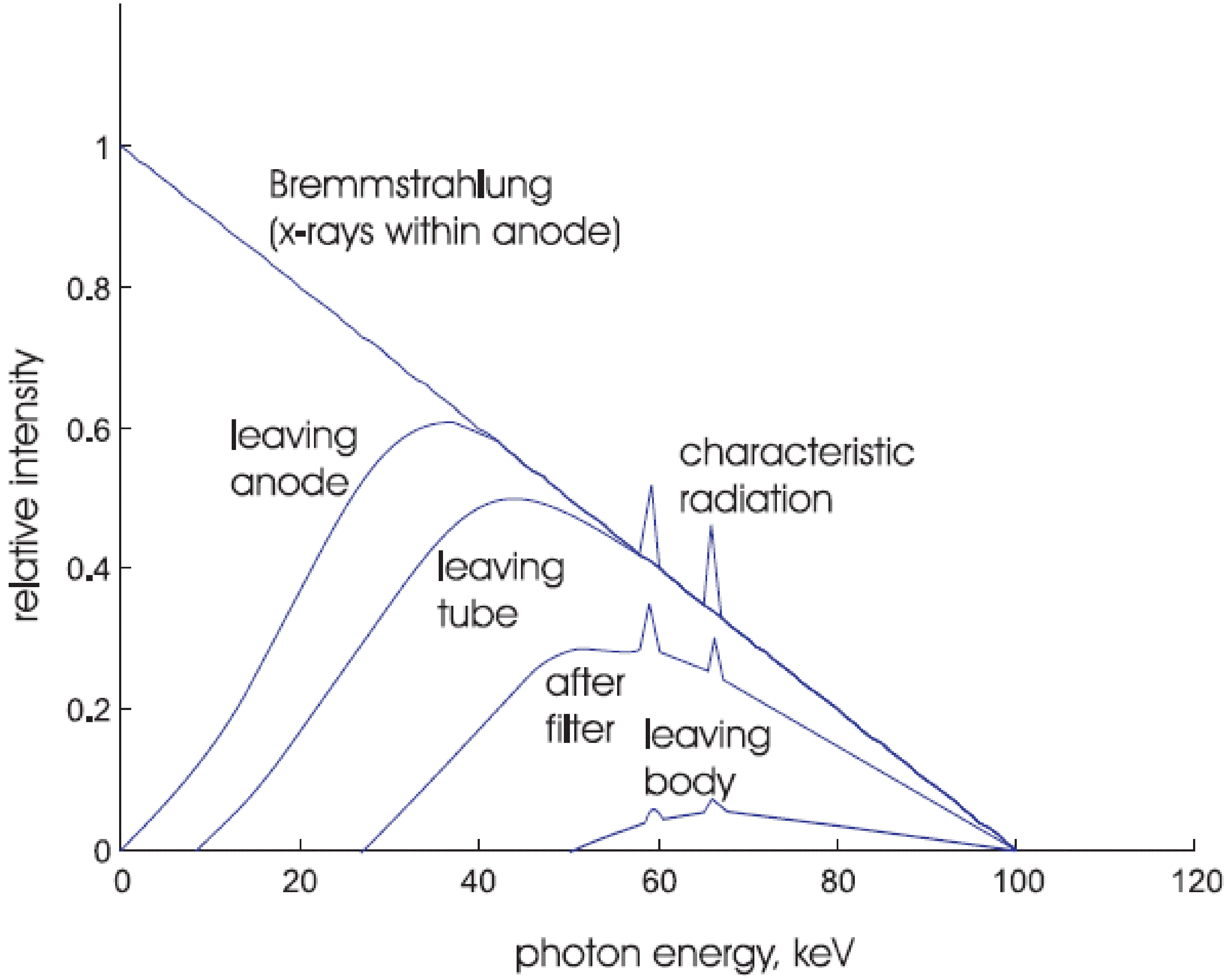
- Filament controls tube current (mA)
- Cathode and focussing cup
- Anode is switched to high potential
  - 30–150 kVp
  - Made of tungsten
  - Bremsstrahlung is 1%
  - Heat is 99%
  - Spins at 3,200–3,600 rpm
- Glass housing; vaccum

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
    - Itube = 1-1000mA
- mA times exposure time yields mAs

mAs measures x-ray exposure

X-Ray Spectra

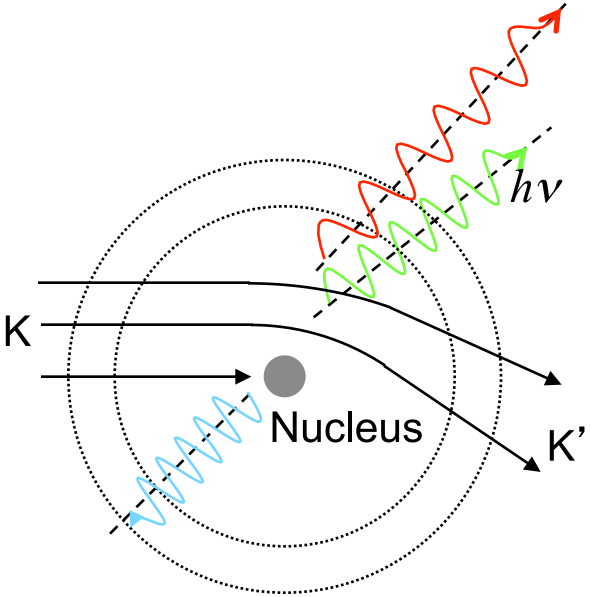


Bremsstrahlung

- *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:

$$E_{p,\max} = h\nu = K_e = eV_{tube}$$

[From Graber, Lecture Note for BMI1-FS05]

Bremsstrahlung intensity

- Overall Bremsstrahlung intensity I:

$$I \propto V_{tube}^2 I_{tube}$$

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

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

$$P_x = k\, Z\, V_{tube}^2 I_{tube} = kZ\, V_{tube} P_{tube} = \eta P_{tube}$$

$$\eta = P_x / P_{tube} \quad = \quad kZ\, V_{tube}: \quad \text{x-ray production efficiency}$$

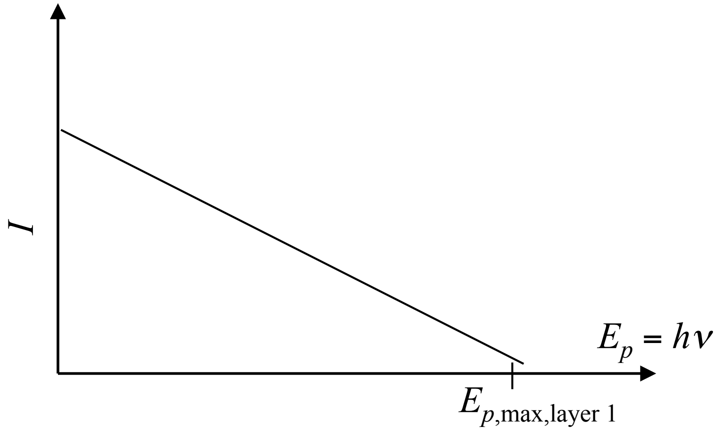
  - Material constant $k = 1.1 \times 10^{-9}$ for Tungsten (Z=74).

[From Graber, Lecture Note for BMI1-FS05]

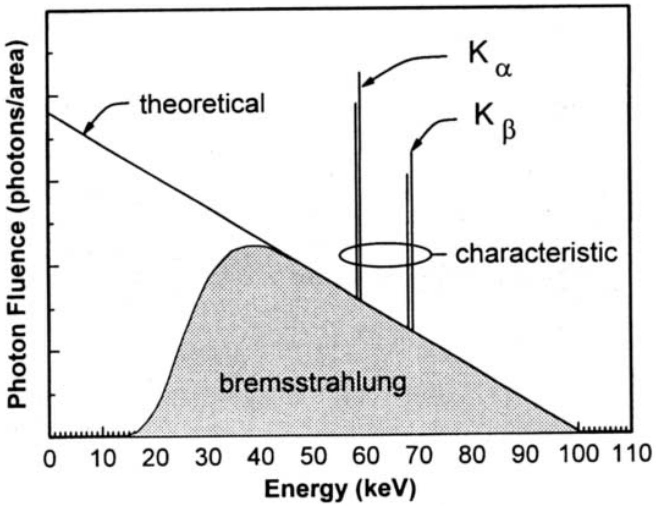
Bremsstrahlung spectrum

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

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



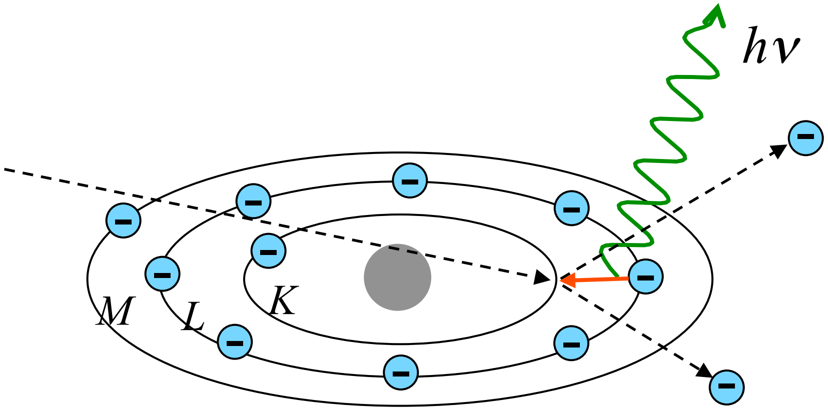
- Actual spectrum deviates from ideal form due to
  - Absorption in window / gas envelope material and absorption in anode
  - Multienergetic electron beam

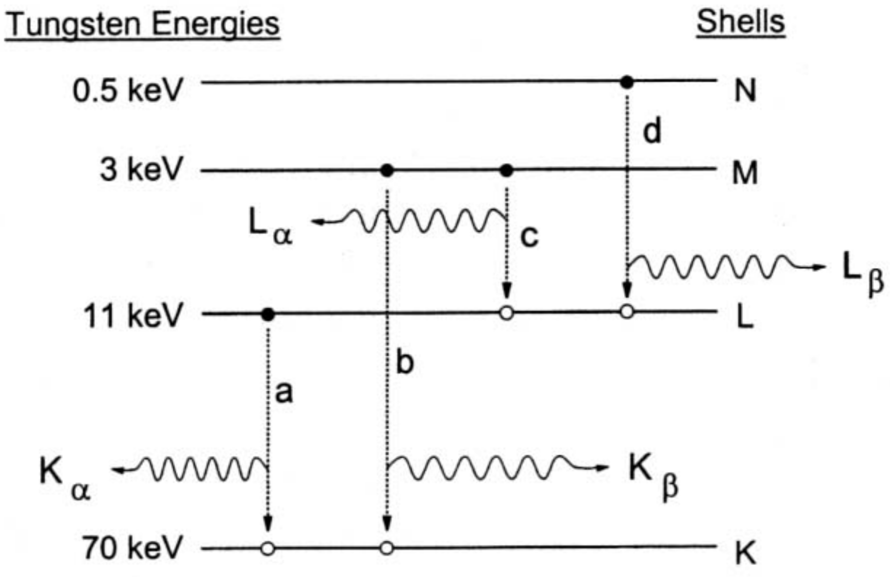


[From Graber, Lecture Note for BMI1-FS05]

Characteristic radiation

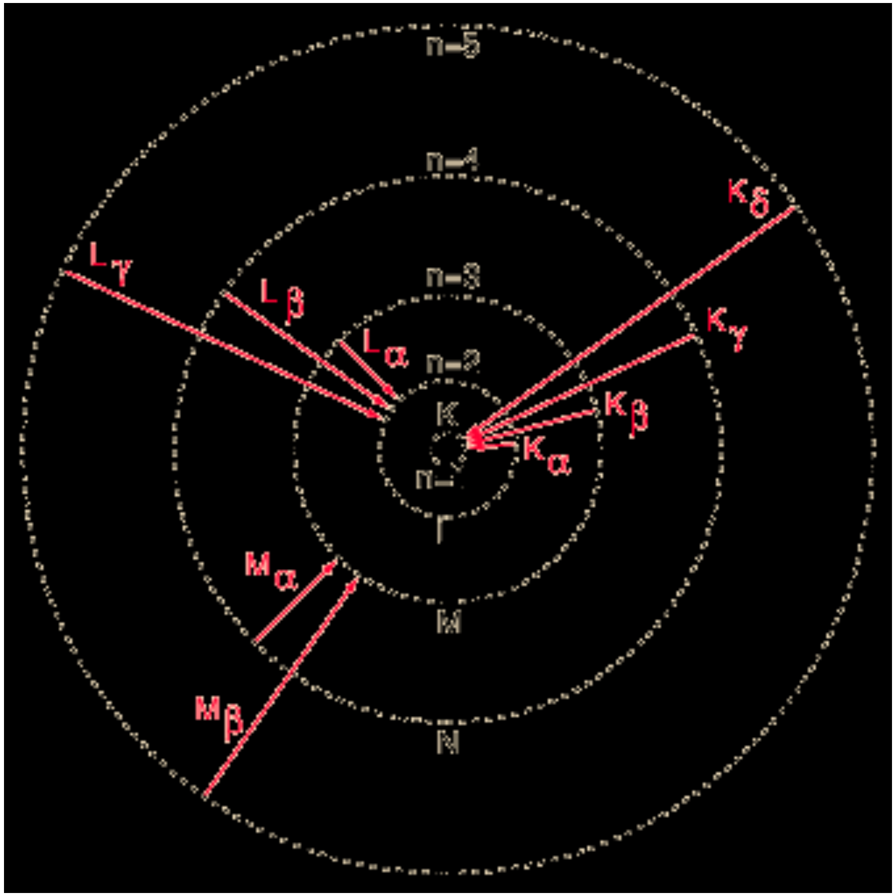
- Narrow lines of intense x-ray at characteristic energies are superimposed on the continuous bremsstrahlung spectrum.
- Caused by removal of inner shell electrons and subsequent filling of hole with electrons from higher shell. The shell-energy difference determines the energy of characteristic rays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





[From Graber, Lecture Note for BMI1-FS05]

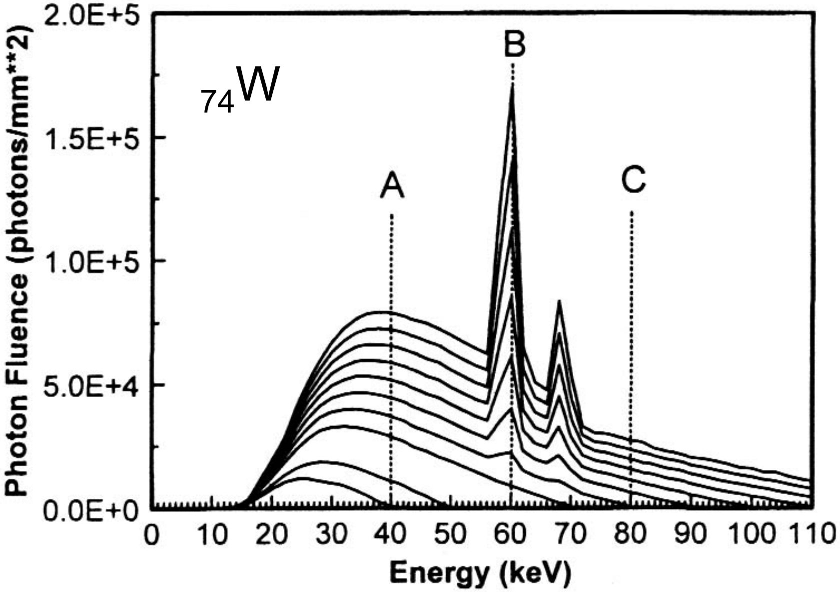
Different types of characteristics rays



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

X-ray spectra

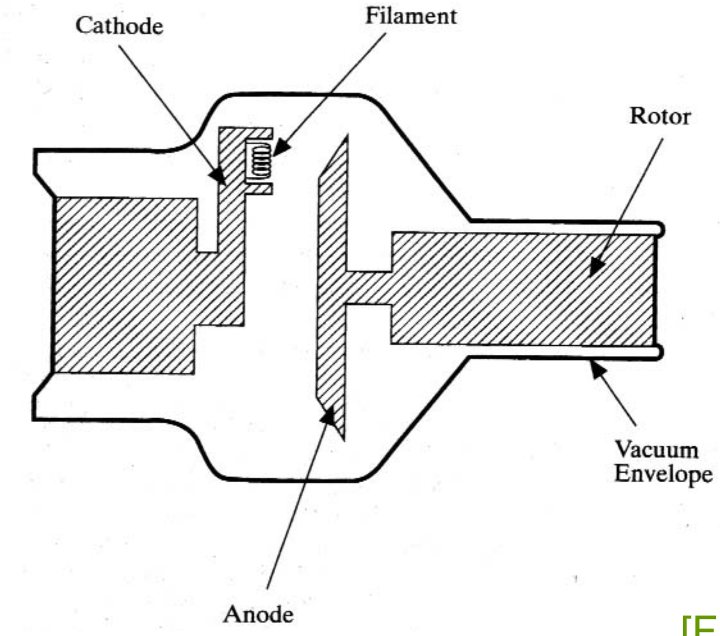
- X-ray for general diagnostic radiology produced at 40 – 150 kVp
- Maximum photon energy: $E_p[\text{keV}] = h\nu_{\max} = e \times \text{kVp}$
- Characteristic radiation occurs only for anode voltages
$$e \times \text{kVp} > I_{K,L,M,\ldots}$$

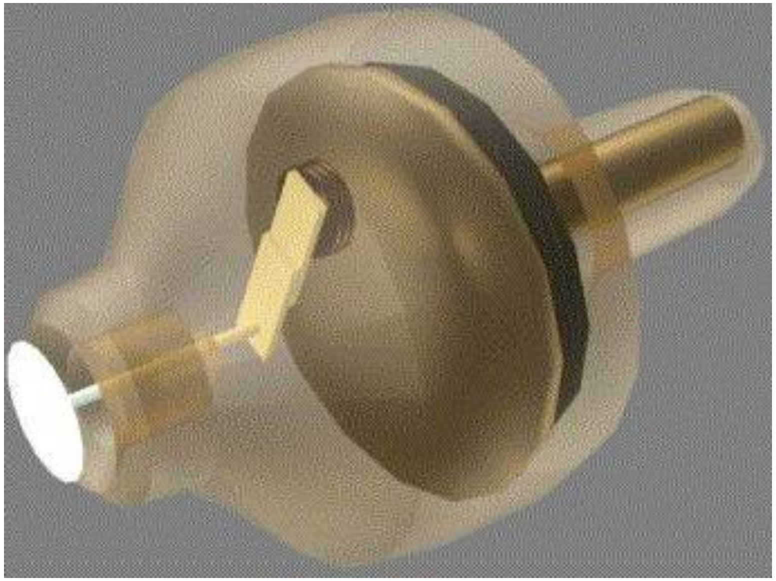


[From Graber, Lecture Note for BMI1-FS05]

X-ray tube design

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





[From Graber, Lecture Note for BMI1-FS05]

Filtration

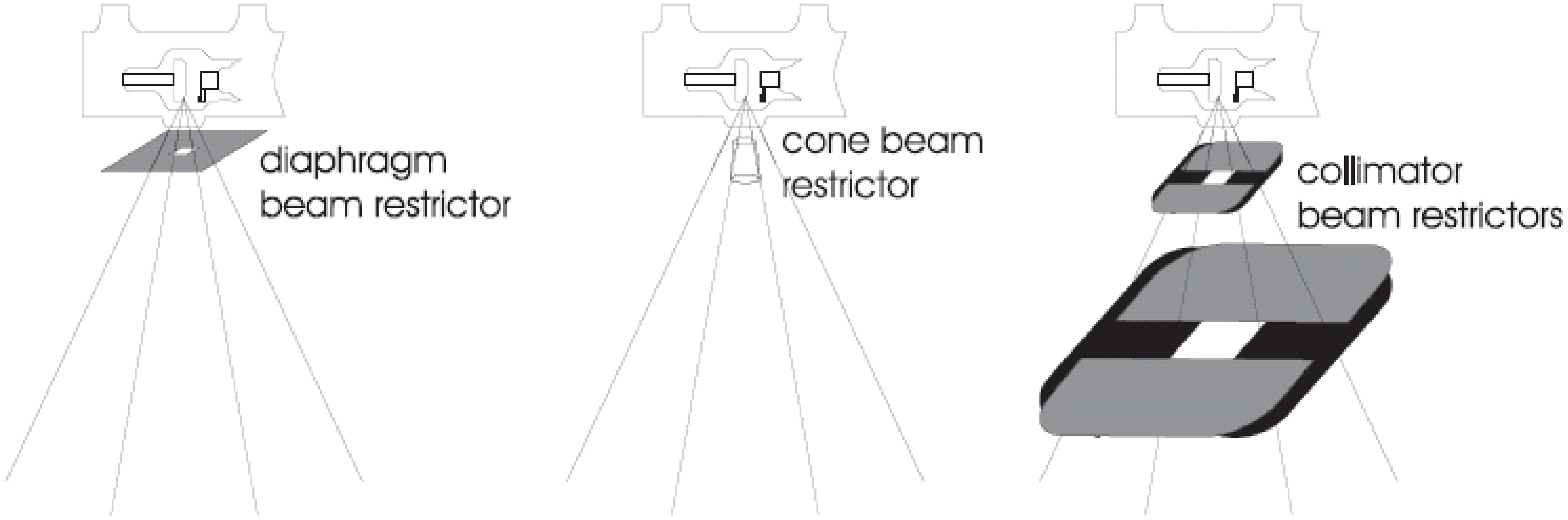
- Low energy x-ray will be absorbed by the body, without providing diagnostic information
- Filtration: Process of absorbing low-energy x-ray photons before they enter the patient

- Inherent filtration
  - Within anode
  - Glass housing
- Added filtration
  - Aluminum
  - Copper/Aluminum

    Note: Cu has 8keV characteristic xrays
  - Measured in mm Al/Eq

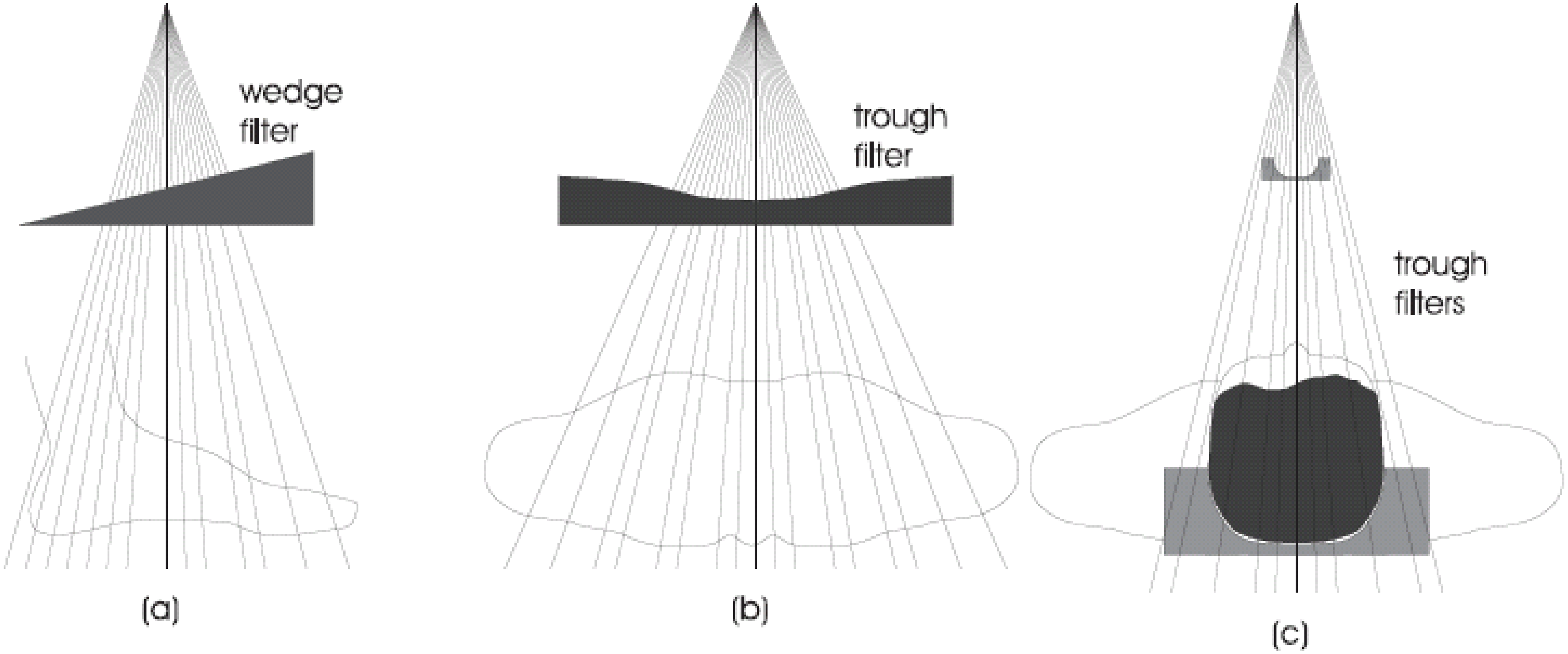
Restriction

- Goal: To direct beam toward desired anatomy



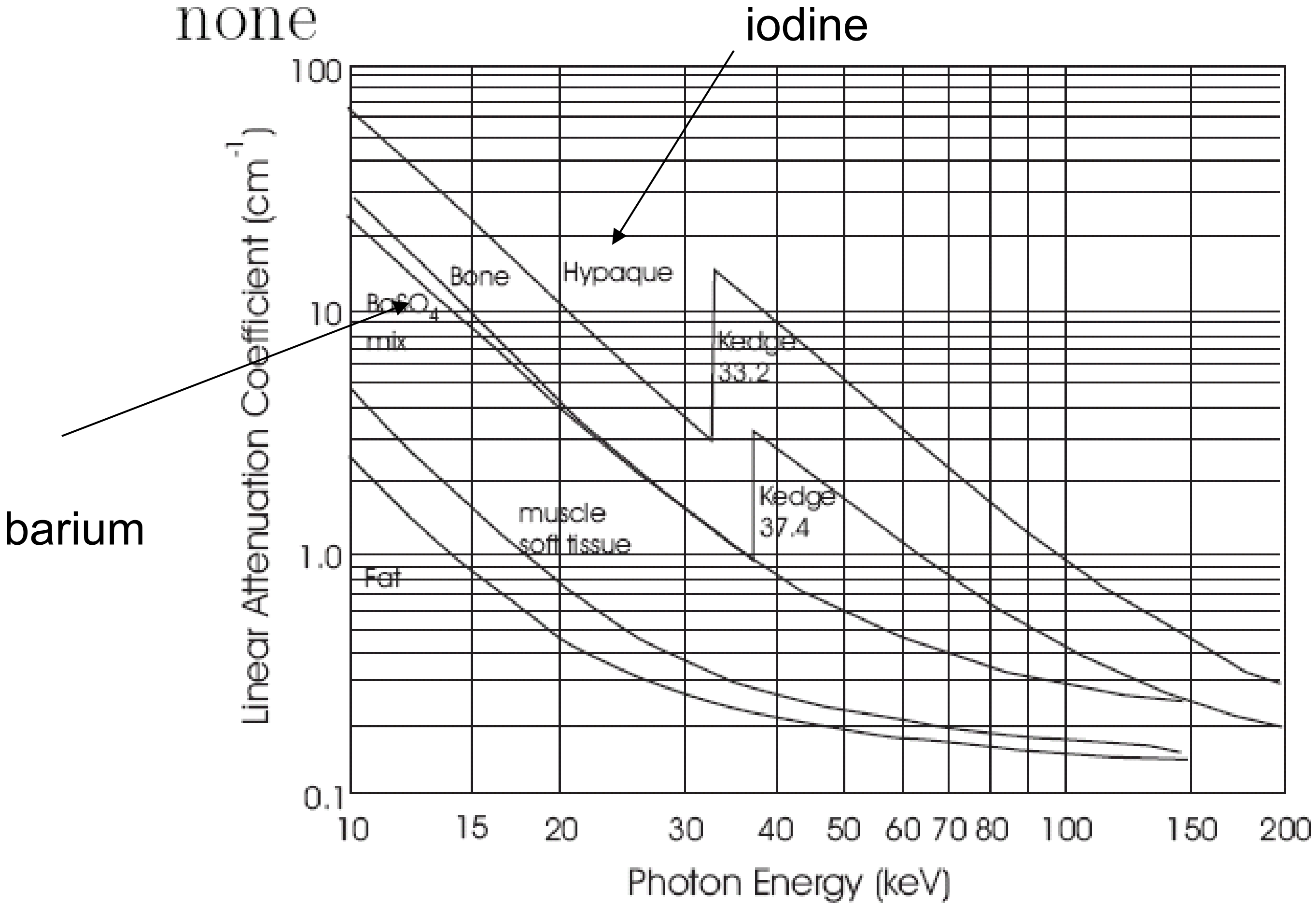
Compensation Filters

- Goal: to even out film exposure



Contrast Agents

- Goal: To create contrast where otherwise none



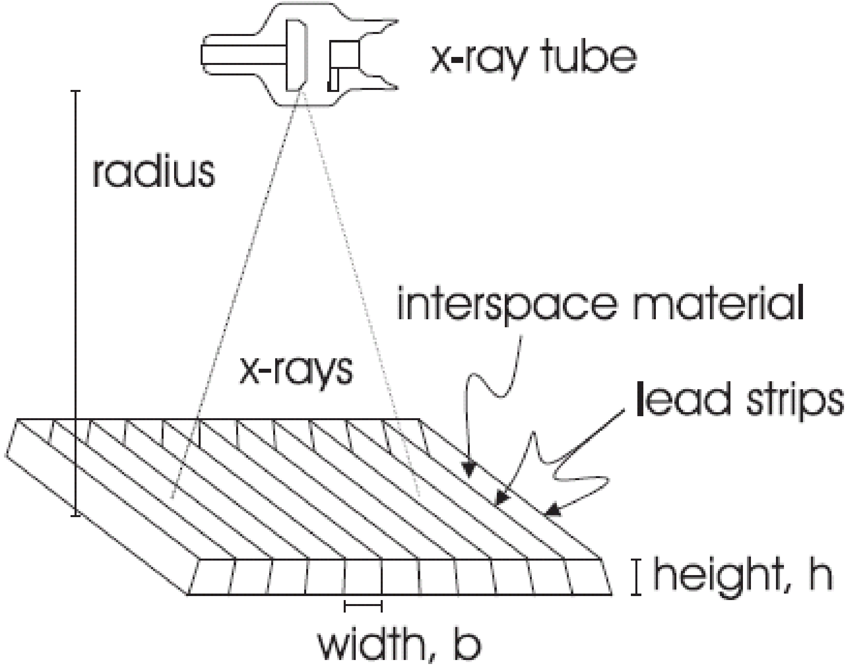
When the x-ray energy exceeds the Kedge (binding energy of K-shell), the mu coefficient is much higher, providing high contrast

- 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: <u>a line!</u>
- 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

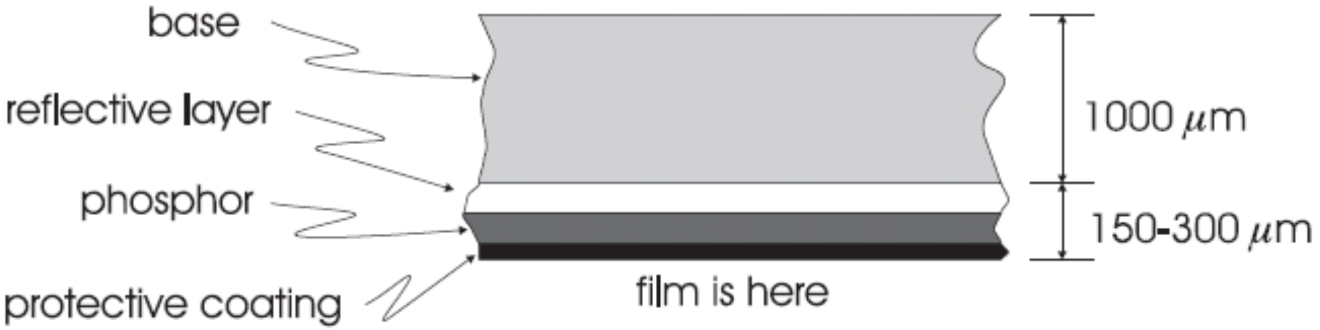
$$\mathrm{GCF} = \frac{\text{mAs w/ grid}}{\text{mAs w/o grid}}$$

  - Typical range $3 < \mathrm{GCF} < 8$
- 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:

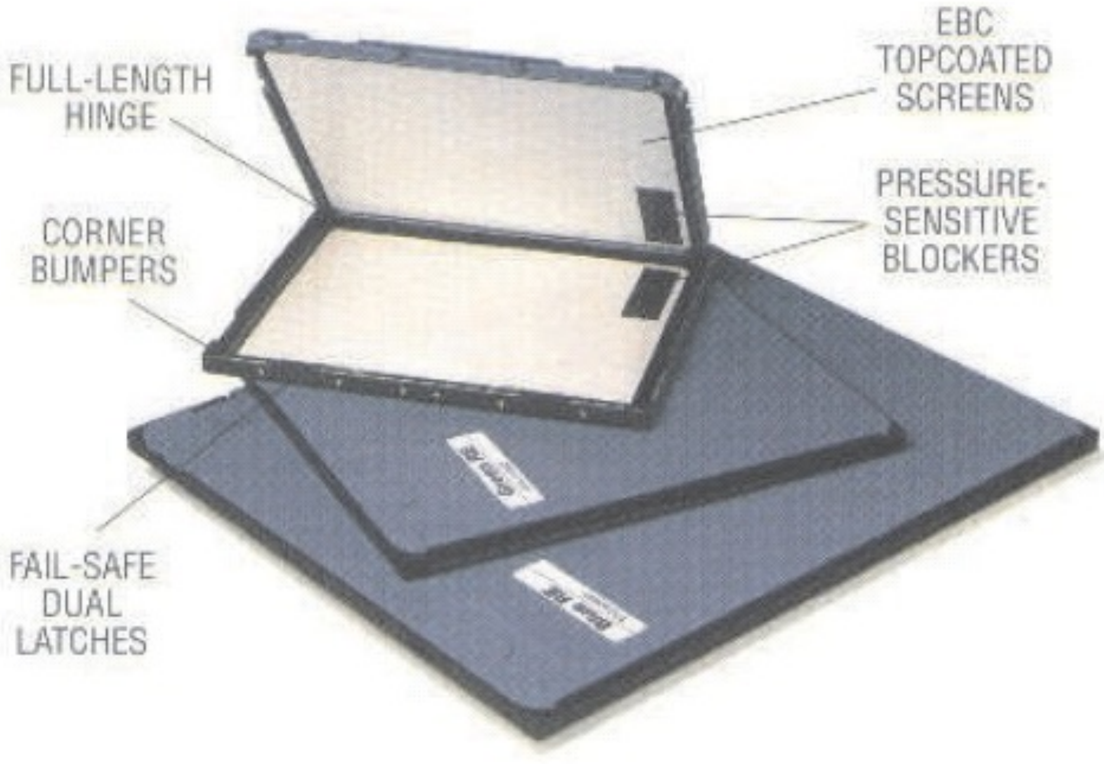


Phosphor:
convert x-ray
photons to light

Reflective layer:
Reflect light back
to film

- Phosphor = calcium tungstate
- Flash of light lasts 1×10^{-10} second
- ~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

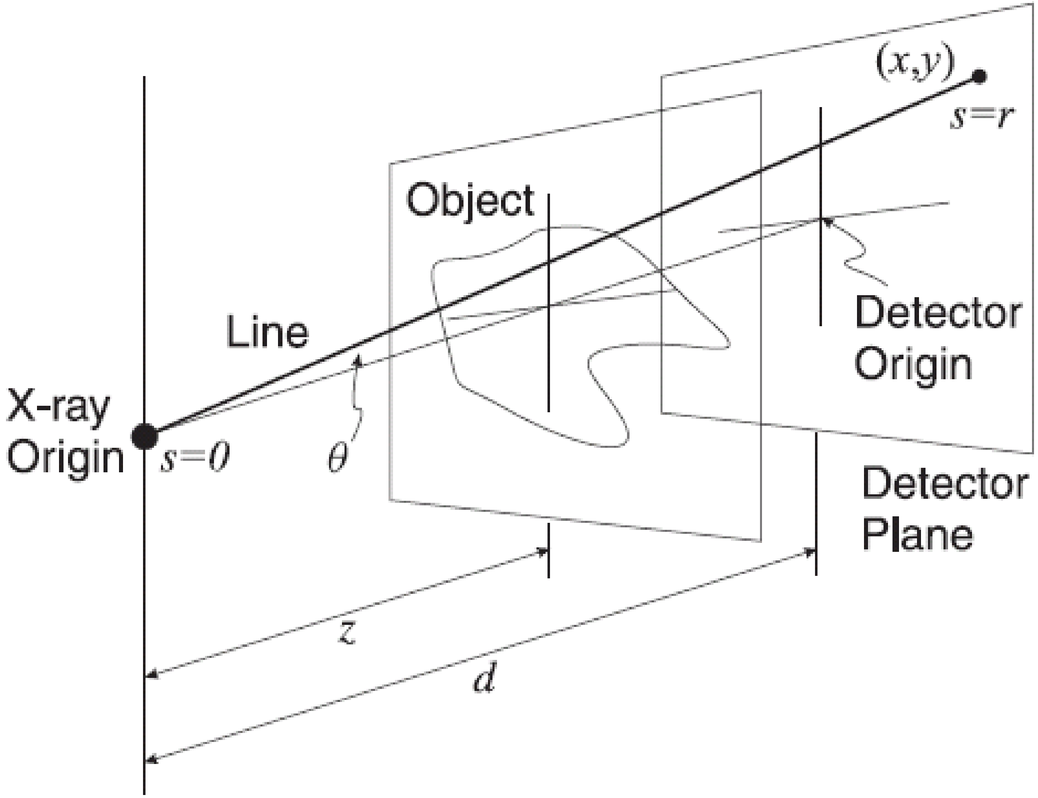
- Damage depends on deposited (= absorbed) energy (intensity × time) per tissue volume
- Threshold: No minimum level is known, above which damage occurs
- Exposure time: Because of recovery, a given dose is less harmful if divided
- Exposed area: The larger the exposed area the greater the damage (collimators, 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

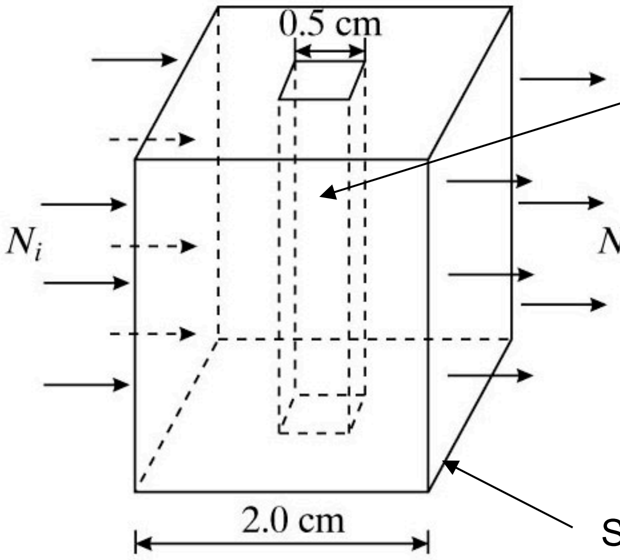
- Basic imaging equation
- Geometric effects
- Extended source
- Film blurring
- Impact of noise and scattering

Basic Imaging Equations



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

Example



Blood vessel μ=0.2

w/contrast μ=20

Soft tissue μ=0.4

w/o contrast :

$$I_b = I_{\min} = I_0 e^{-(0.4*2.0)};$$

$$I_o = I_{\max} = I_0 e^{-(0.4*1.5+0.2*0.5)}$$

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

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

w/ contrast :

$$I_b = I_{\max} = I_0 e^{-(0.4*2.0)};$$

$$I_o = I_{\min} = I_0 e^{-(0.4*1.5+20*0.5)}$$

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

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

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?

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
- θ is angle between $(0,0)$ and (x,y)

Inverse Square Law

- Net flux of photons decrease as $1/r^2$.
  Therefore

$$I_0 = \frac{I_S}{4\pi d^2} \qquad I_r = \frac{I_S}{4\pi r^2}$$

- Eliminate source intensity I_S

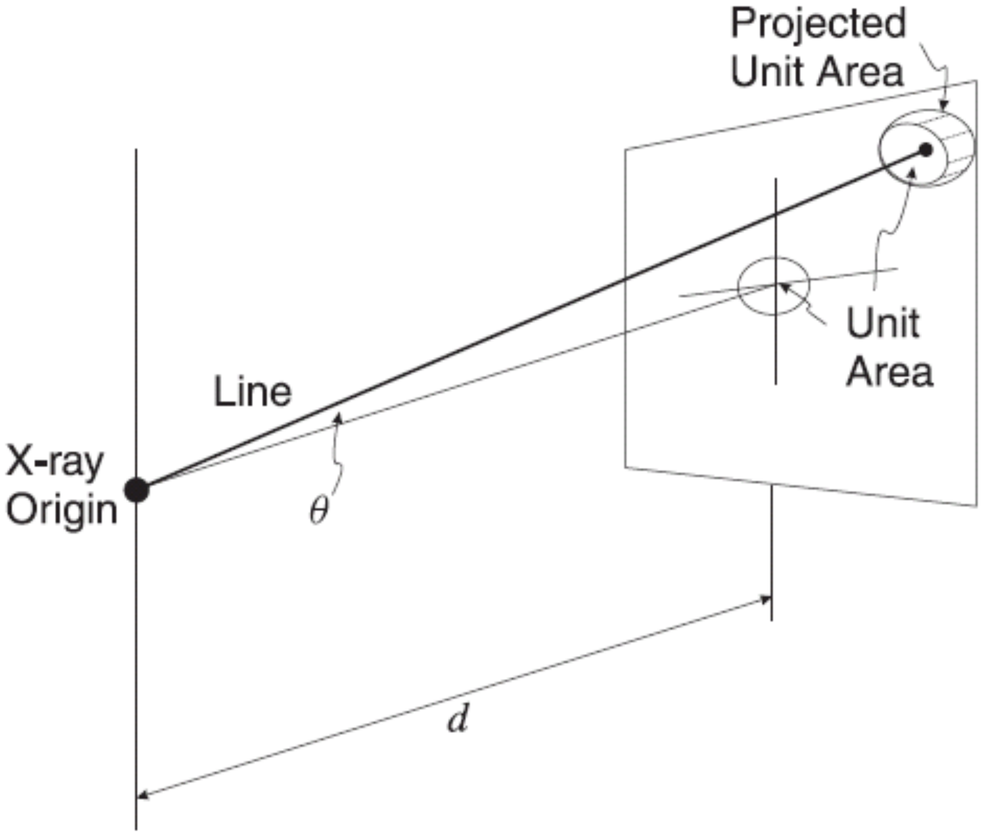
$$I_r = I_0 \frac{d^2}{r^2}$$

- 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 θ w/o considering the oblique effect discussed in the next page

Obliquity



I_0 should be replaced by I_r

- Intensity is

$$I_d = I_0 \cos\theta$$

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

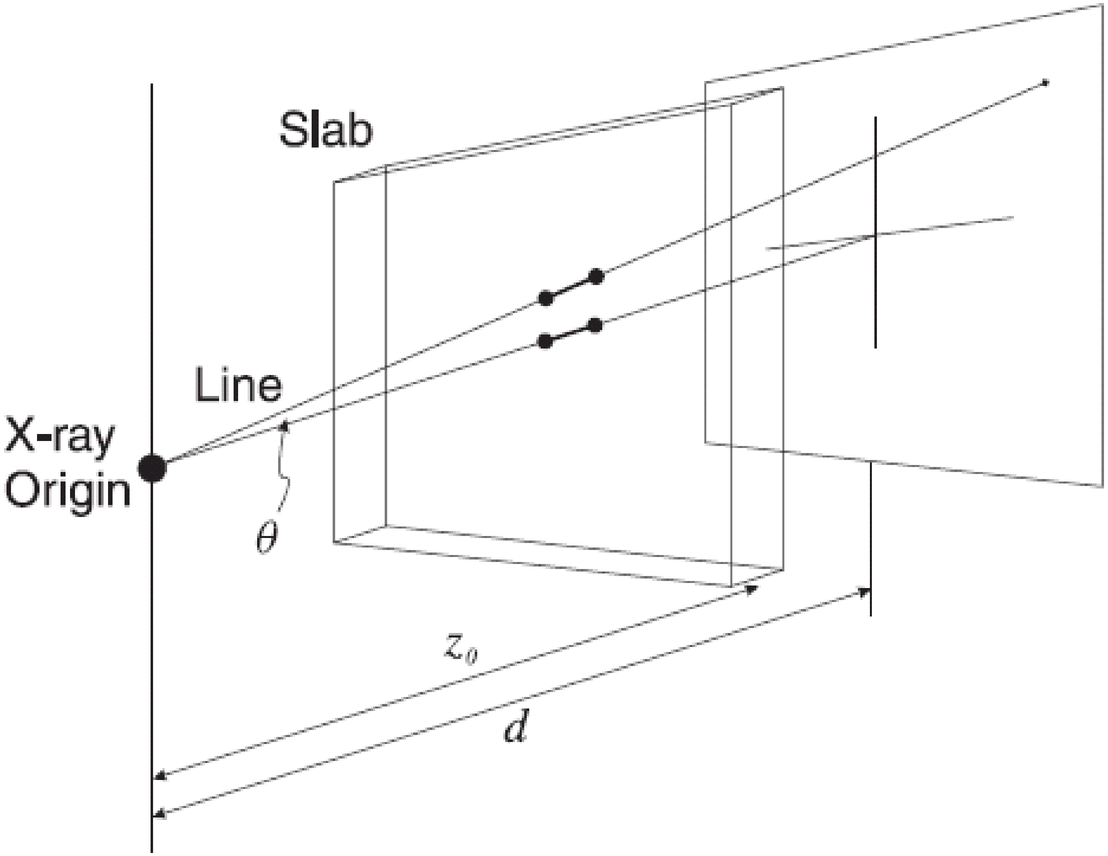
−> θ is small

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

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

- Including inverse square law and obliquity:

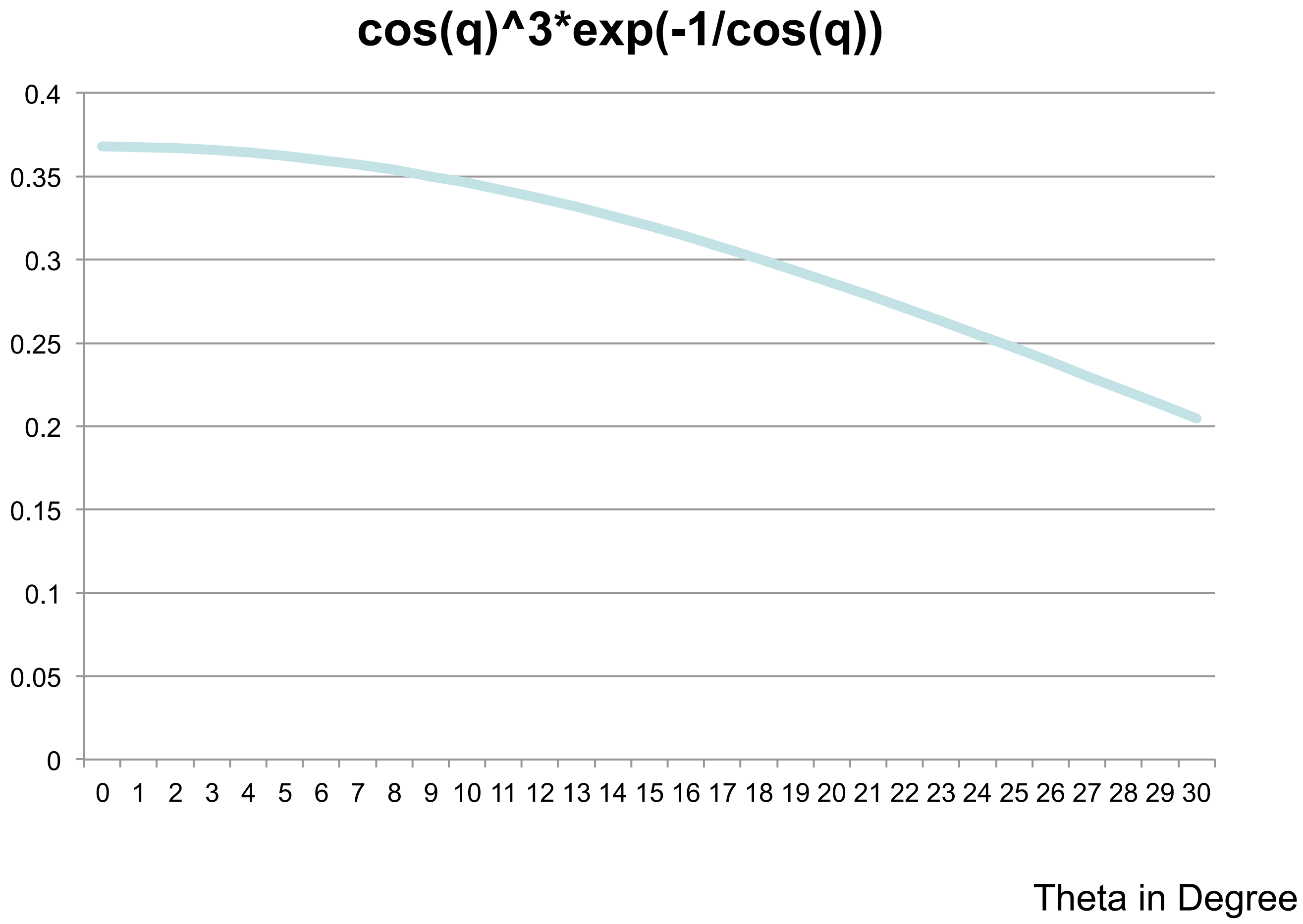
$$I_d(x, y) = I_i \cos^3\theta \exp\{-\mu L / \cos\theta\}$$

- If $d \approx r$ <u>all effects can be ignored</u>

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

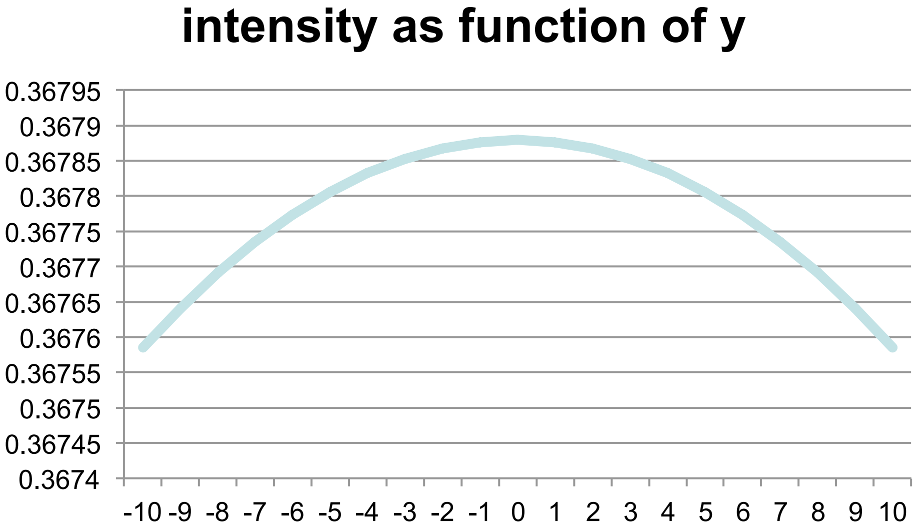
Illustrate the received intensity as function of y or x or \theta

Received Signal as a Function of Theta



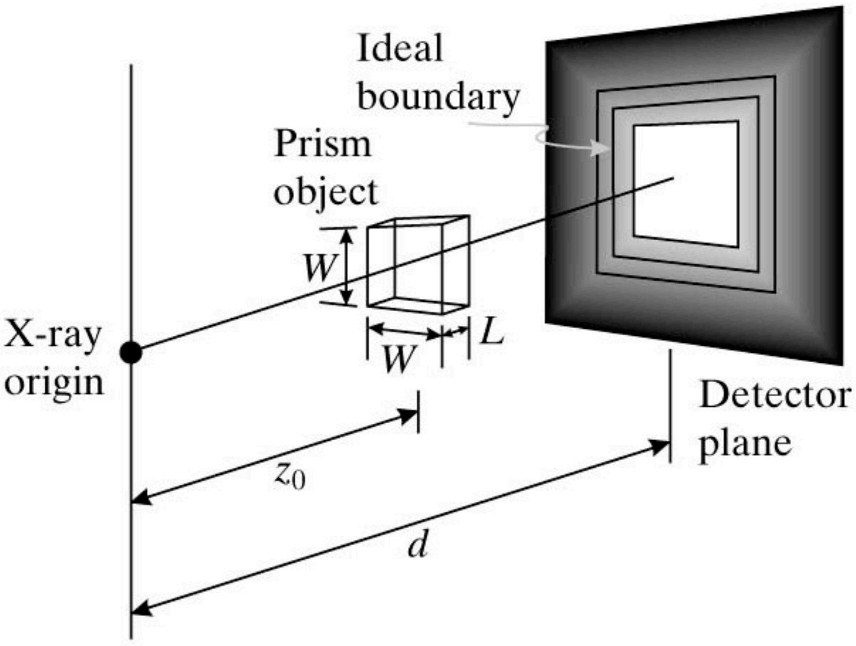
This plot assumes mu*L=1, e.g. mu=1/cm, L=1cm.

- How does it vary as a function of Y when x=0 (vertical axis of the detector plain)?
- cos(q) = d/r=d/sqrt(d^2+y^2)/
- Assuming d=5m, y= -10cm to 10cm (q from 0 to 1.14 degree)
- Vary small relative change in the range 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 *I0*, and the object has a constant linear attenuation coefficient m. (Example 5.4 in textbook)

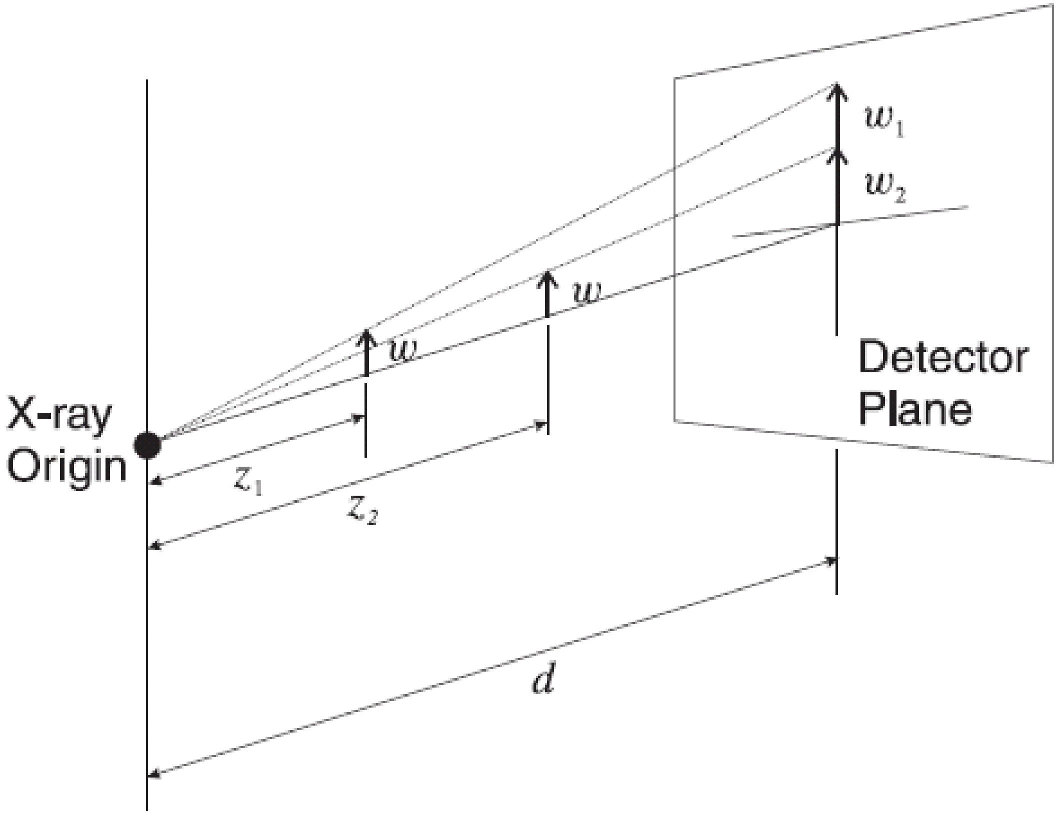


Solution

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

Objects Magnification

- Size on detector depends on distance from source



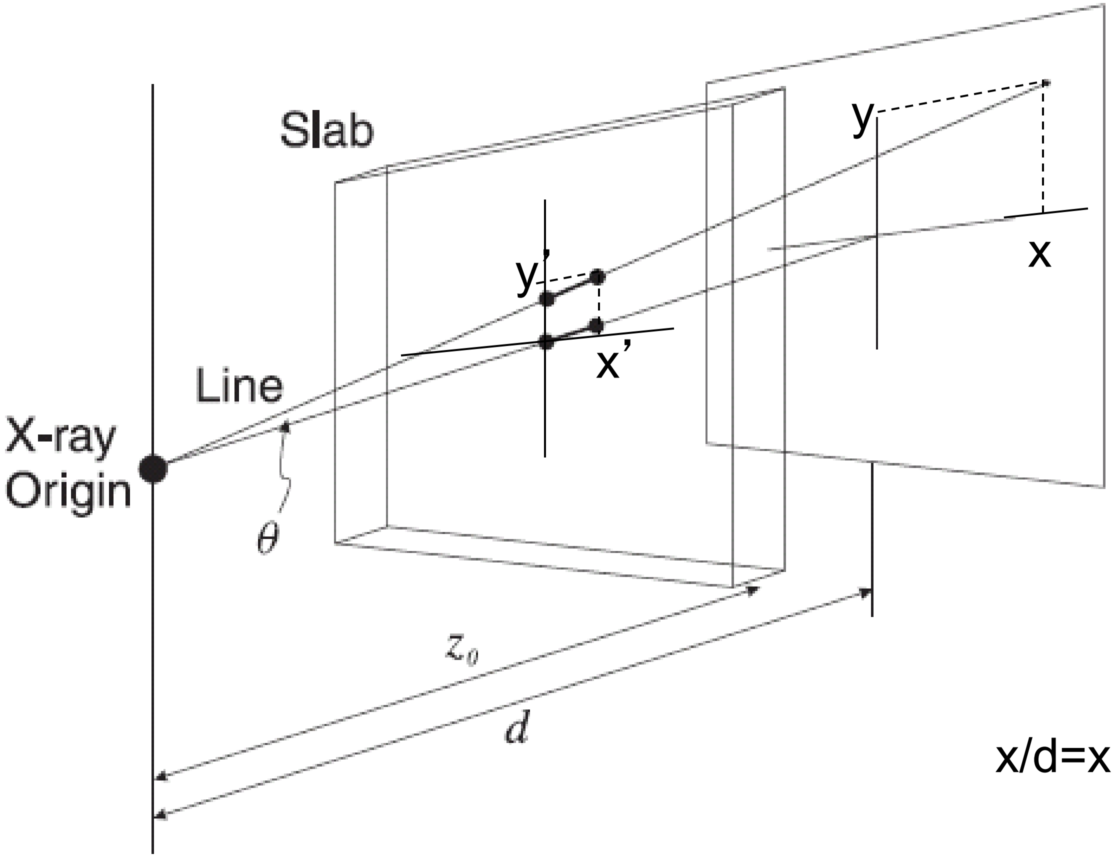
$$w_z = w\frac{d}{z}$$

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 -> x' =x z/d = x /M(z)

- Let "transmittivity" be

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

- On detector, intensity is

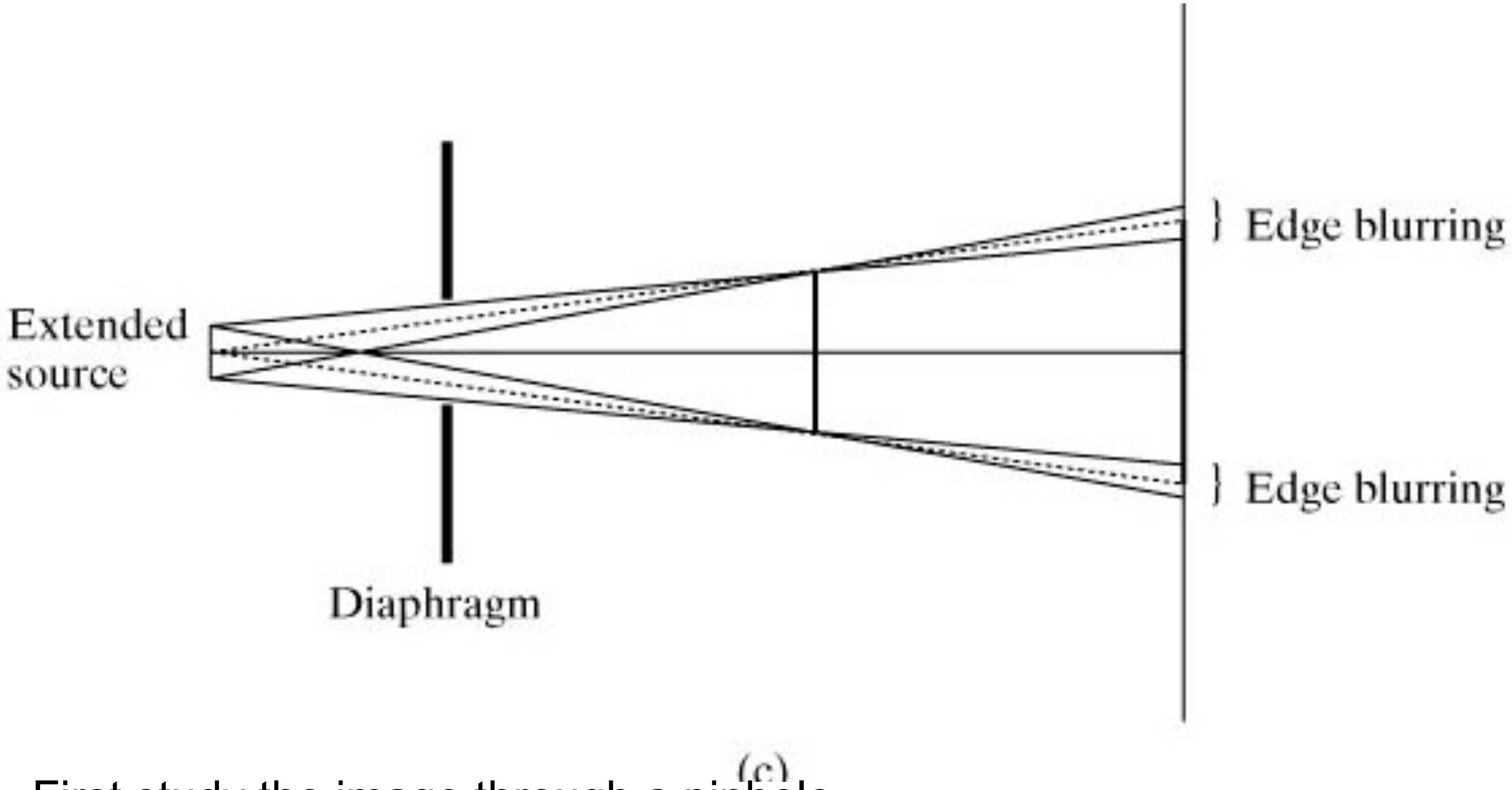
$$I_d(x, y) = I_0 \cos^3\theta \, t_z\left(\frac{x}{M(z)}, \frac{y}{M(z)}\right)$$

- After substitution

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

I0=Is / (4 pi d^2)

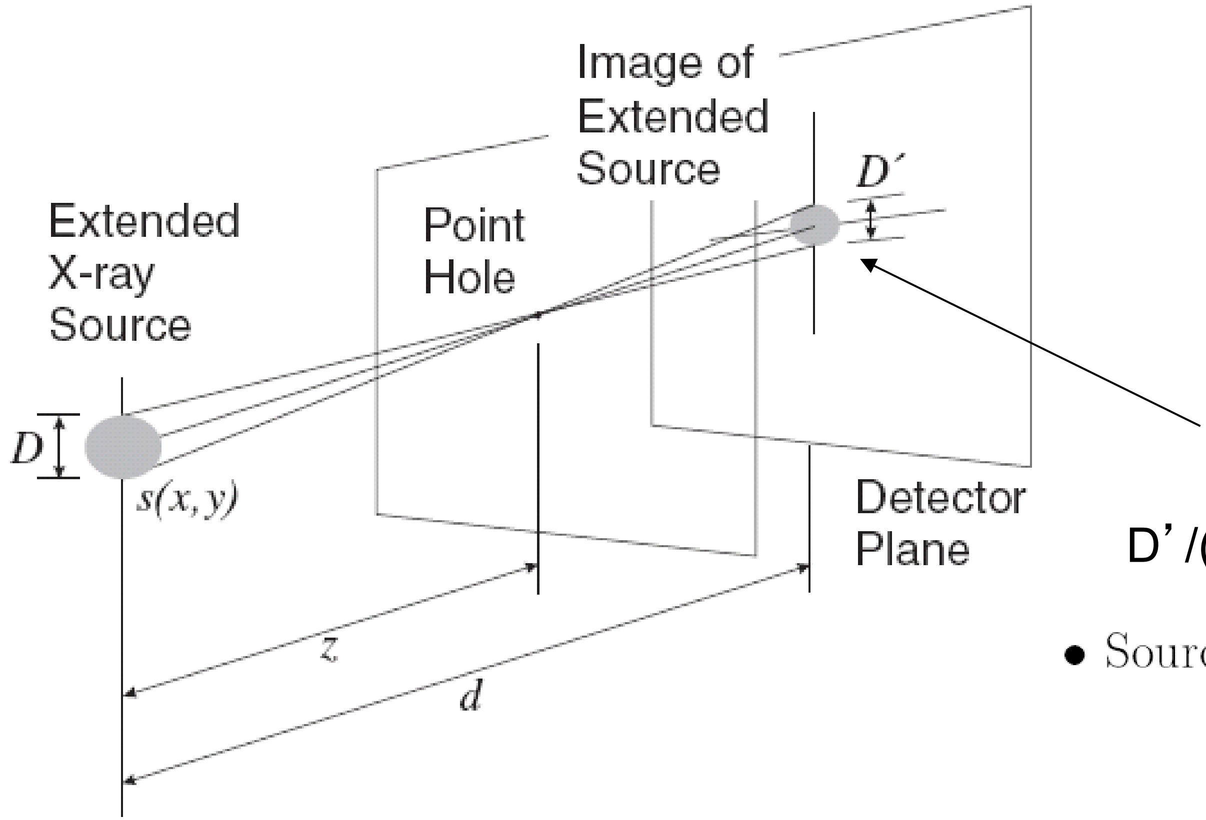
Blurring Due to Extended Source



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 -> 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

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

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 $t_z(x',y')$
  - *The received signal due to transmitivity 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') tz(x',y') dx' dy'*
- The overall effect can be captured through linear con

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

Note: m 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

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

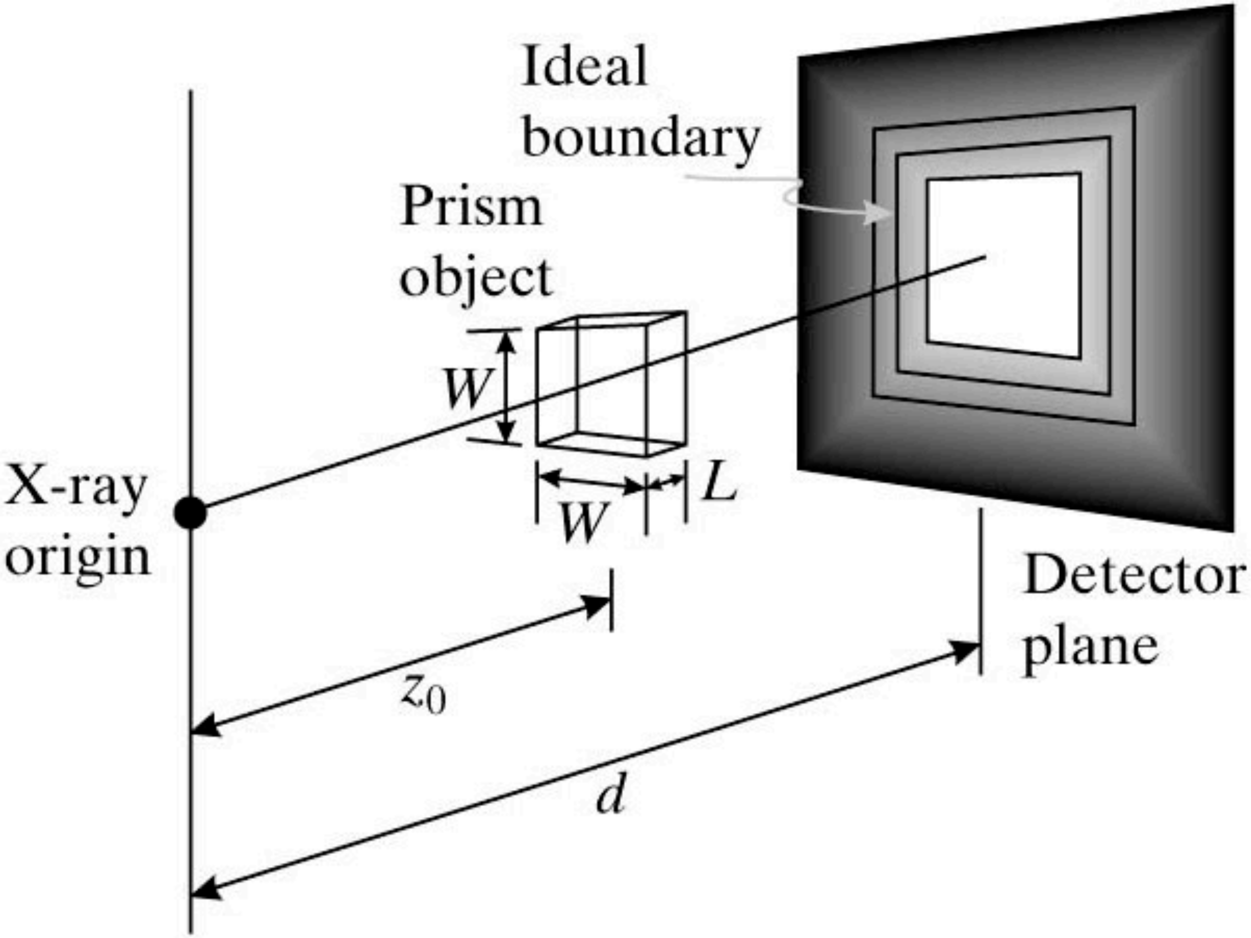
  The resulting detector image is a square with width 2W but with a blurred edge with blurring width D

- For d=3z, M=d/Z=3, m=1-M=-2

  Tz(x/M,y/M): a square with width 3W

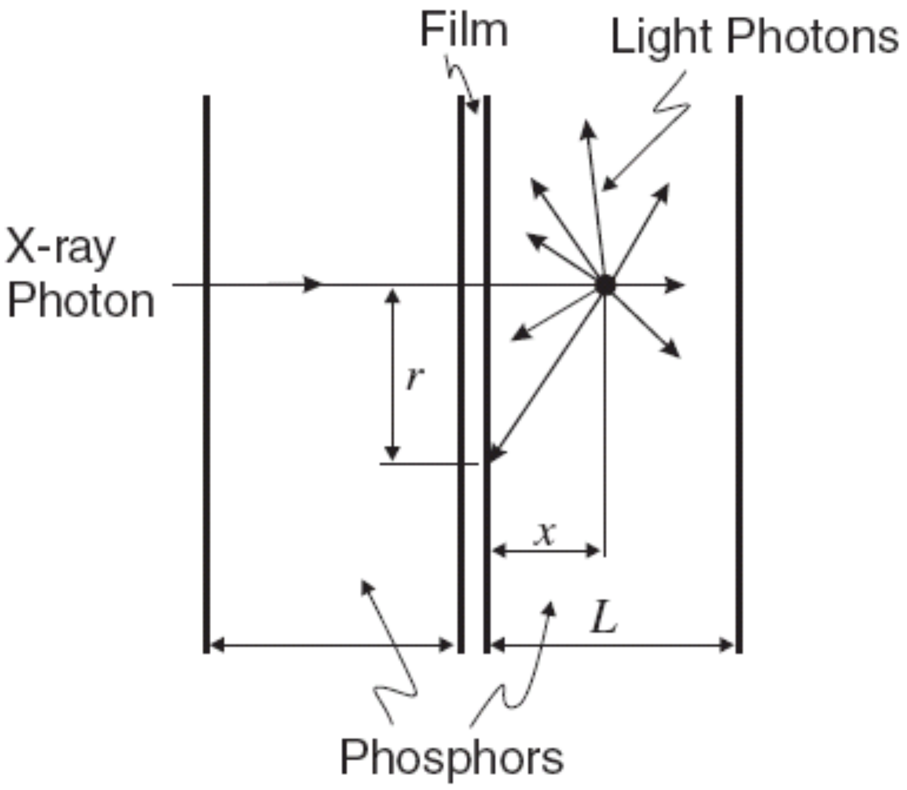
  S(x/m,y/m): a disk with diameter 2D

  The resulting detector image is a square with width 3W but with a blurred edge with blurring width 2D



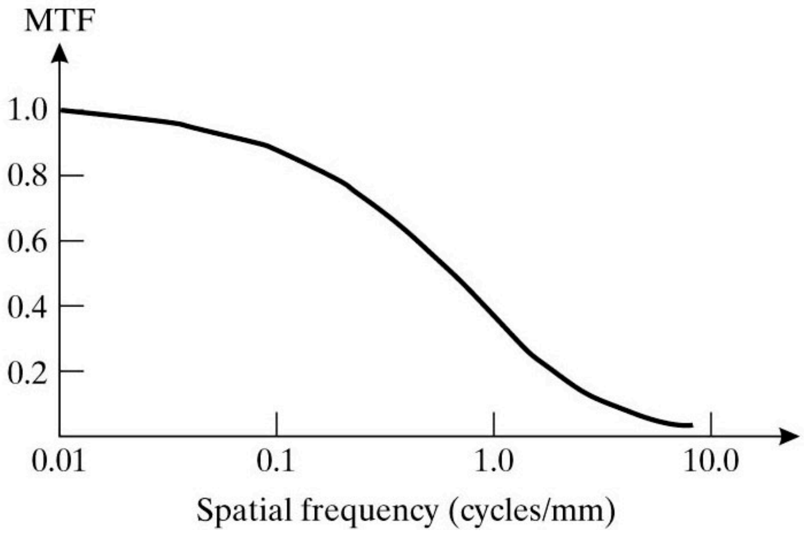
For the previous example, L is very small, but the source has diameter D, blurring is due to the source diameter being non-zero

Film Screen Blurring



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

$$\boxed{\begin{aligned} I_d(x,y) &= \cos^3\theta \frac{1}{4\pi d^2 m^2} s\left(\frac{x}{m},\frac{y}{m}\right) \\ &\quad * t_z\left(\frac{x}{M},\frac{y}{M}\right) * h(x,y) \end{aligned}}$$

- 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

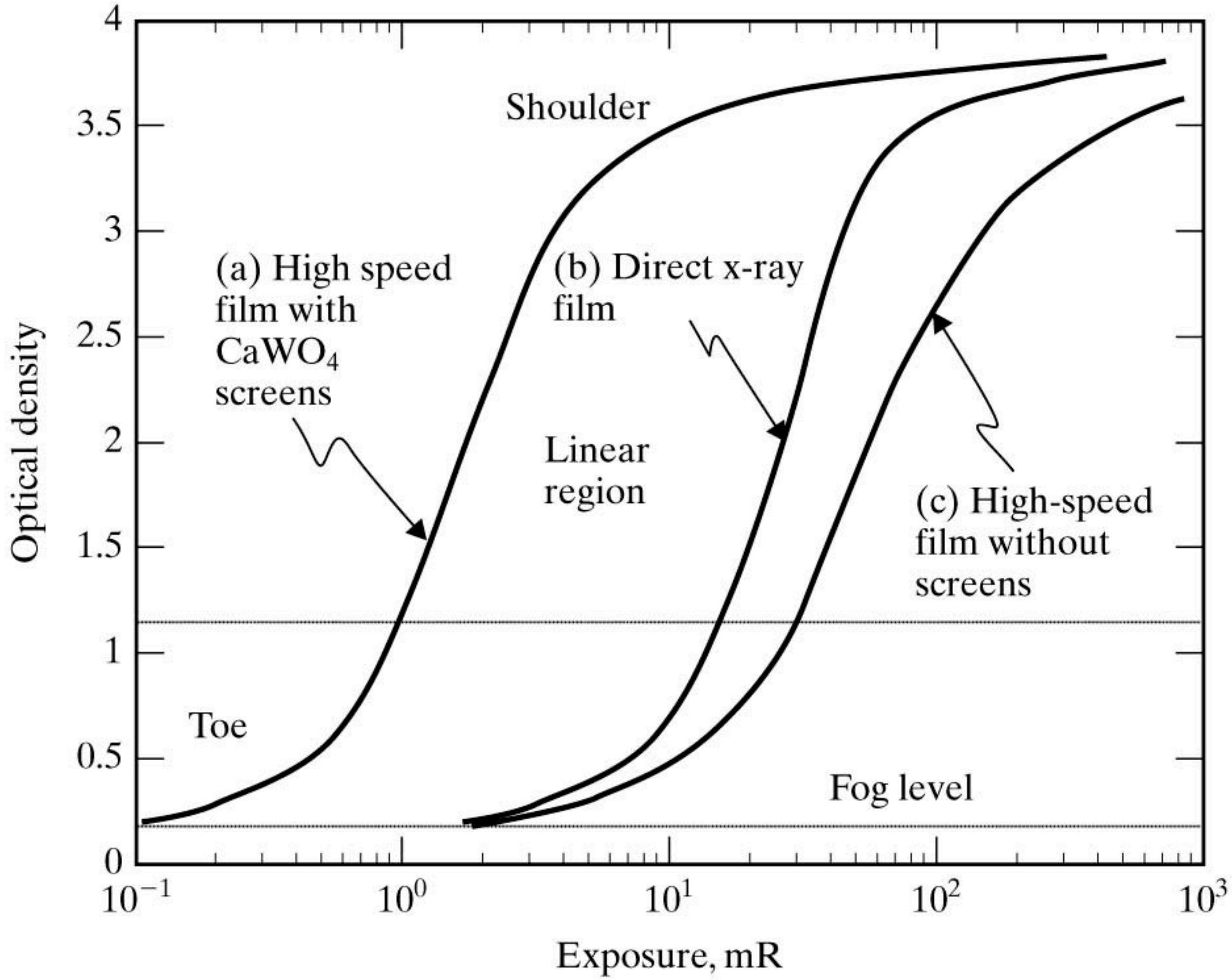
- X-ray exposure yields optical density

$$D = \Gamma \log_{10} \frac{X}{X_0}$$

- Γ 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

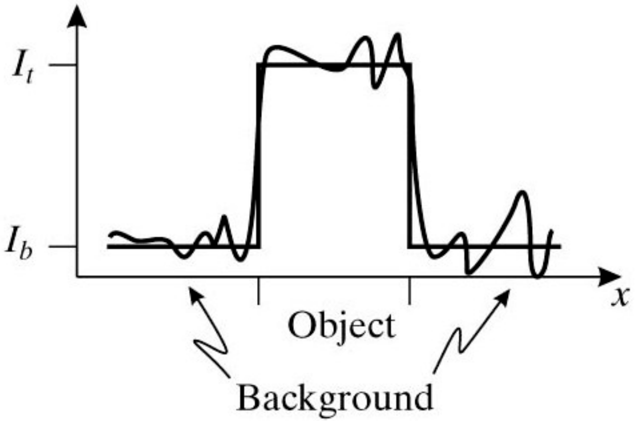
$$D = 1 + \text{ fog level}$$

The H&D Curve


Shoulder
(a) High speed film with $CaWO_4$ screens
(b) Direct x-ray film
Linear region
(c) High-speed film without screens
Toe
Fog level
Optical density
4
3.5
3
2.5
2
1.5
1
0.5
0
10^{-1}
10^{0}
10^{1}
10^{2}
10^{3}
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: σ_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)= (a^k / k!) e^{-a}
  - Variance = mean = a

- Let N_b denotes the average number of photons per burst per area
- Let hv 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

$$\sigma_b^2 = N_b \left(\frac{hv}{A\Delta t} \right)^2$$

- The SNR is $\mathrm{SNR} = C\sqrt{N_b}$

If X is an RVwith mean η_x, variance σ_x^2

$Y = aX$ is a RV with mean $\eta_y = a\eta_x$, variance $\sigma_y^2 = a^2\sigma_x^2$

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

Detective Quantum Efficiency

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

$$\mathrm{DQE} = \left(\frac{\mathrm{SNR}_{\mathrm{out}}}{\mathrm{SNR}_{\mathrm{in}}}\right)^2$$

- 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)

$$\mathrm{SNR}_{\mathrm{in}} = \mathrm{mean}\Big/\sqrt{\mathrm{variance}} = \sqrt{10000} = 100$$

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

$$\mathrm{SNR}_{\mathrm{out}} = \mathrm{mean}\Big/\sqrt{\mathrm{variance}} = 8000\Big/\sqrt{40000} = 8000\Big/200 = 40$$

$$\mathrm{DQE} = \left(\frac{\mathrm{SNR}_{\mathrm{out}}}{\mathrm{SNR}_{\mathrm{in}}}\right)^2 = 0.16$$

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:

target intensity: I_t

background intensity: I_b

$$\text{contrast } C = \frac{I_t - I_b}{I_b}$$

$$\text{SNR} = C\frac{I_b}{\sigma_b} = C\sqrt{N_b}$$

W/ scattering:

target intensity: $I_t + I_s$

background intensity: $I_b + I_s$

$$\text{contrast } C' = \frac{I_t - I_b}{I_b + I_s} = \frac{I_b}{I_b + I_s}C = \frac{C}{1 + {I_s}/{I_b}}$$

$$\text{SNR}' = C\frac{I_b}{\sigma_b'} = C\frac{N_b}{\sqrt{N_b + N_s}} = C\frac{\sqrt{N_b}}{\sqrt{1 + N_s / N_b}} = \text{SNR}\frac{1}{\sqrt{1 + I_s / I_b}}$$

Medical Applications

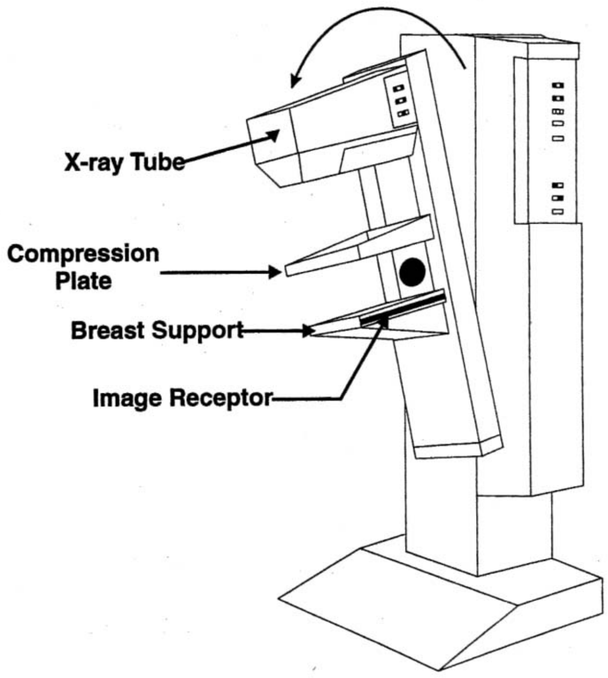
- Orthopedic
- Chest
- Abdomen
- Mammography
- Angiography

[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

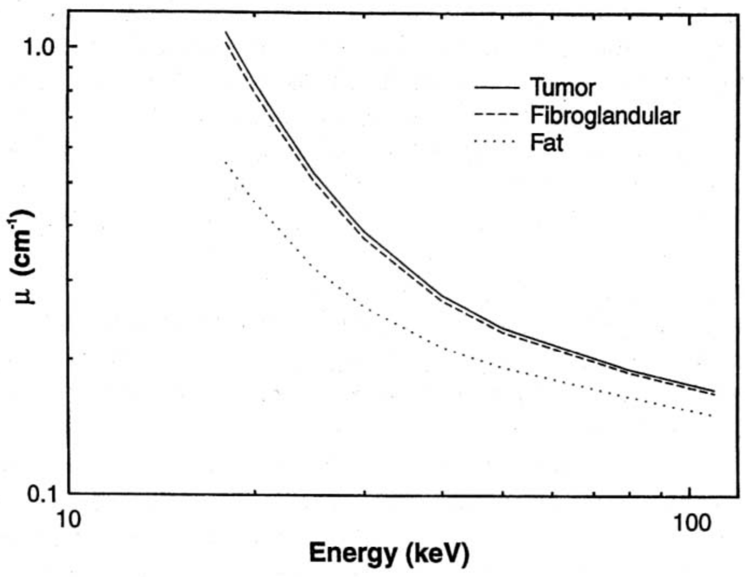


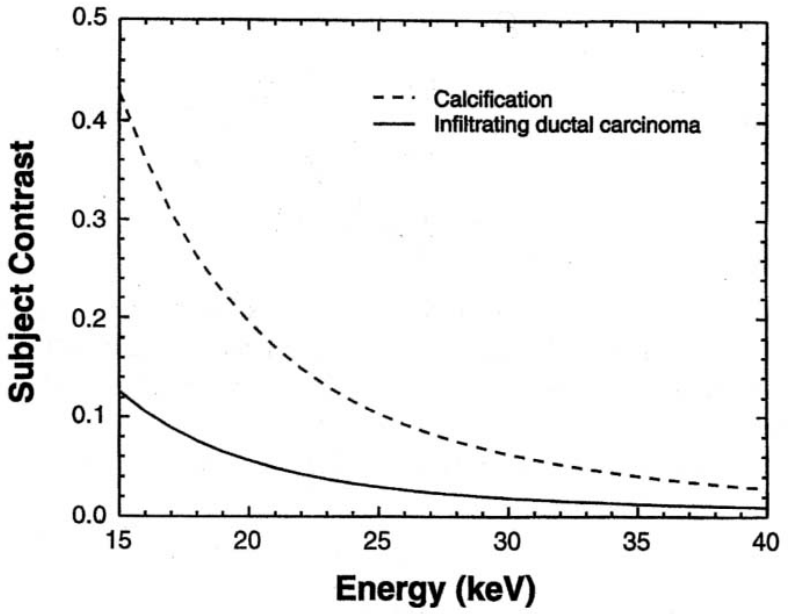
- ⇒ Need for good image contrast of various tissue types.
- Simple x-ray shadowgram from a quasi-point source.

[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), fibroglandular, tumor).
- Ideal energy distribution of X-ray should be below 20 for average size breast, slightly higher for denser breast

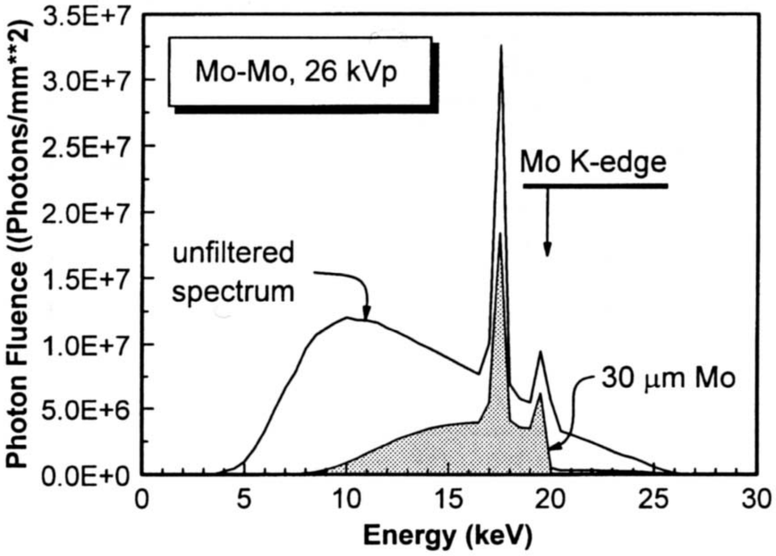




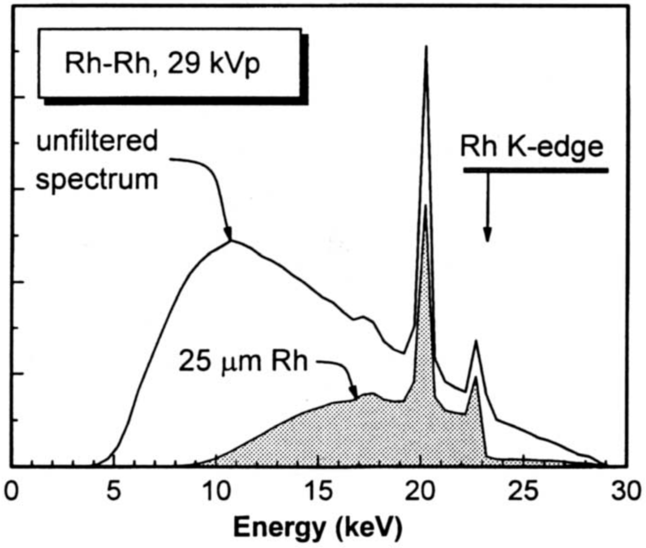
[From Graber, Lecture Note for BMI1-FS05]

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



Target Mo, Filter Mo

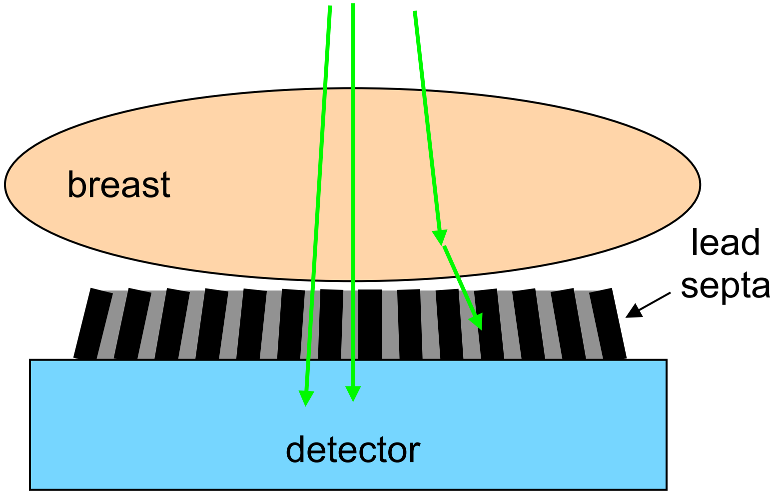


Target Rh, Filter Rh

[From Graber, Lecture Note for BMI1-FS05]

Anti-scatter grid

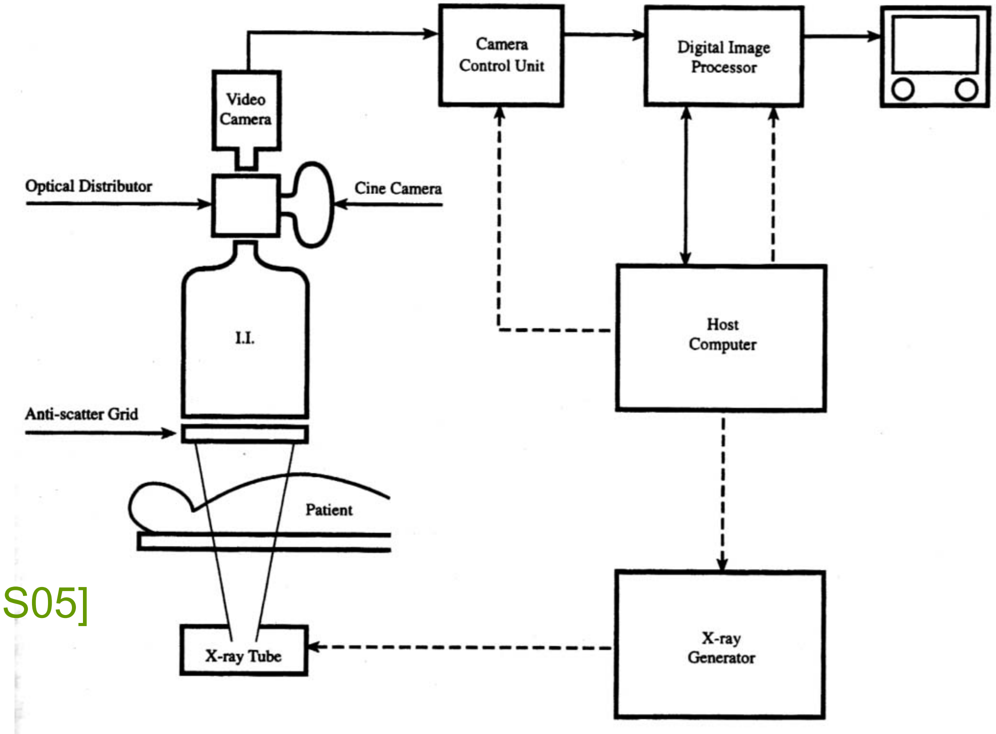
- Significant Compton interaction for low E_p (37-50% of all photons).
- Linear grid: Lead septa + interspace material. Septa focused toward source. Grid ratio ~ 3.5-5:1. Only scatter correction in one dimension. Scatter-to-primary (SPR) reduction factor ~5
- Recently crossed grid introduced
- Grids 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/cm^3
- 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 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)= $(I_{max}-I_{min})/(I_{max}+I_{min})$.
  - P5.22

Homework (added problem)

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 *I0*. For simplicity, assume the slab is infinitely long in the *y* direction.