Questions

- I know the reasons why everything is moving towards digital systems, but based on image quality alone, which is better for these systems, film or digital?
- Not sure how to interpret the left illustration on slide 25. Can you explain?
- Regarding to Voltage determining the X-ray energy Kvp, what is the unit Kvp is equivalent to typical voltage unit?

Email questions to jackie24@uw.edu by Friday April 26

The subject line should be "Phys 428 Lecture 4 Question"

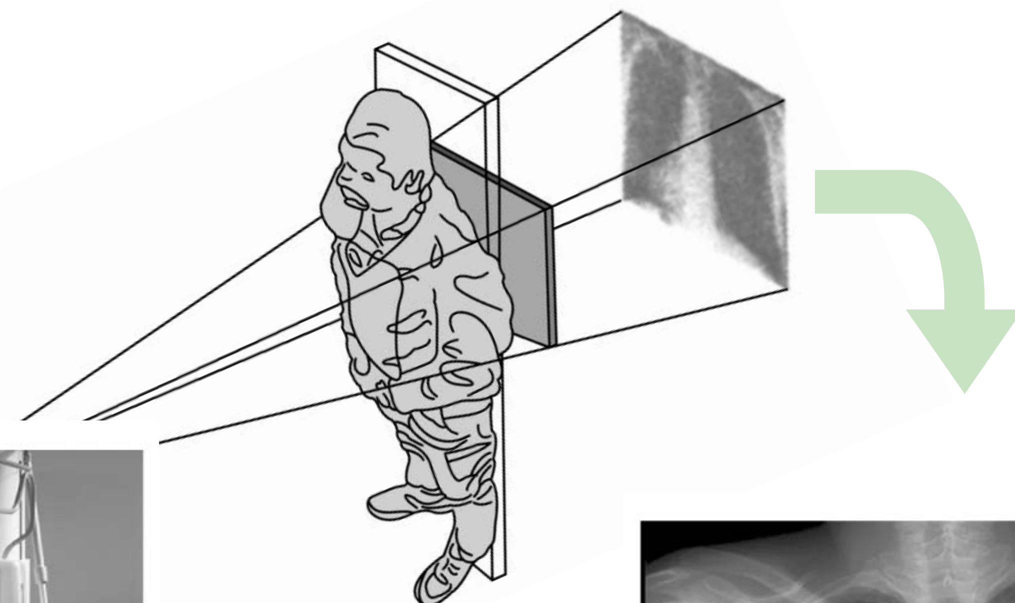
Class Project

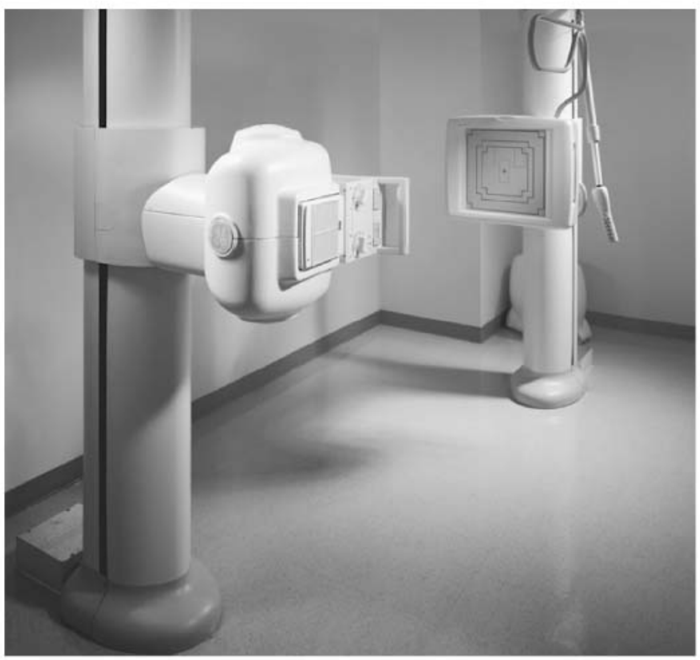
- Pick:
  - An imaging modality covered in class
  - A disease or disease and treatment
- Review:
  - what is the biology of the imaging
  - what is the physics of the imaging
  - what are the competing imaging (and non-imaging) methods
  - what is the relative cost effectiveness of your imaging modality for this disease?
- Form groups (or let me know) by Friday April 26
- 1 page outline Friday May 3 (20%)
- Background summary Friday May 10 (15%)

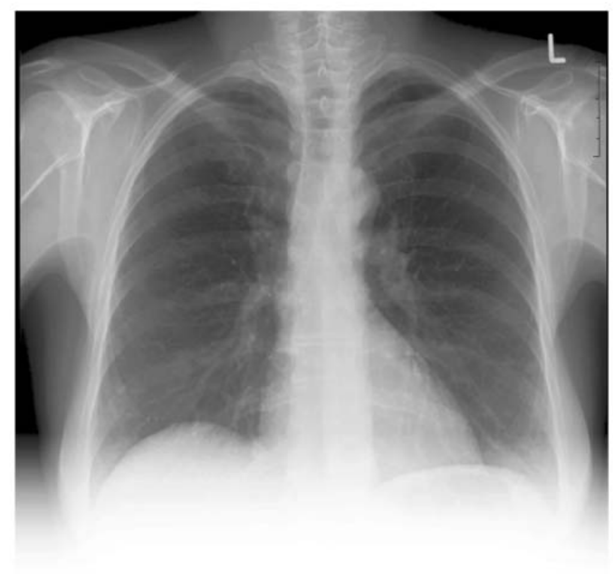
  (what background material you will use & capsule summaries)
- Rough draft Friday May 17 (15%)
- Final version Friday May 31 (30%)
- Presentation / slides Friday June 7 (10%)
- Presentation Tuesday June 11 (10%)

X-ray Computed Tomography

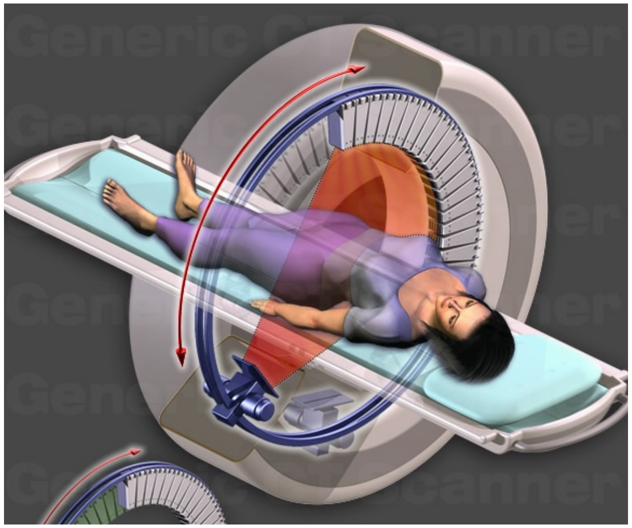
Types of Images: Projection Imaging



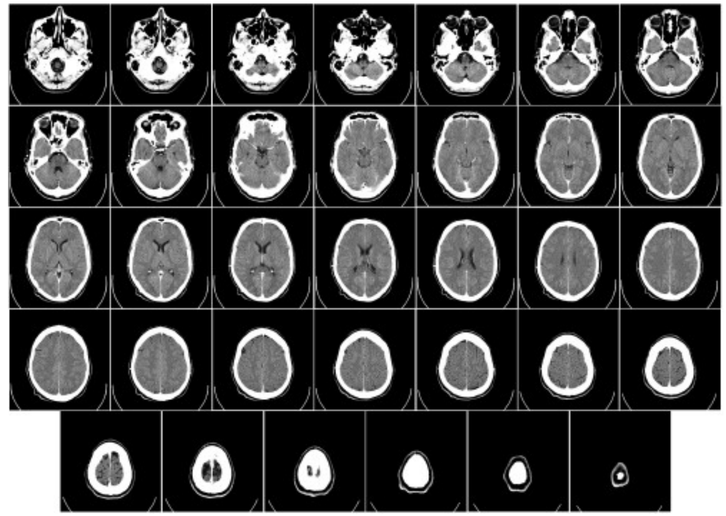




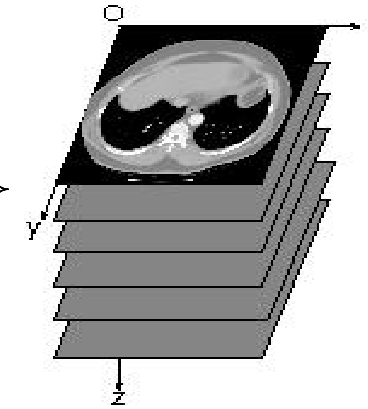
Types of Images: Tomography Imaging



tomographic acquisition

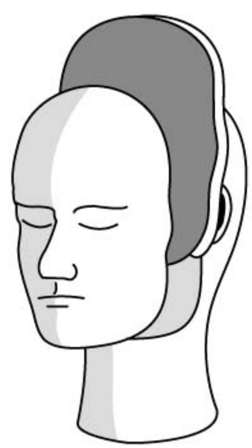


reconstruction of multiple images



form image volume

transaxial or axial view

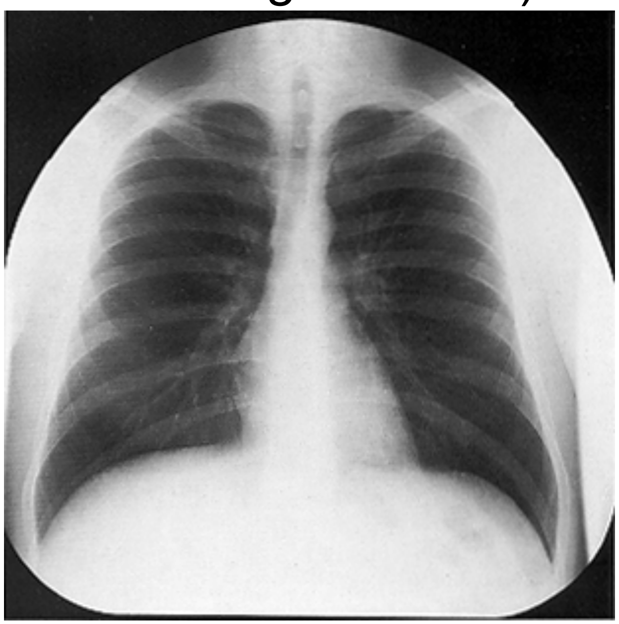


coronal view

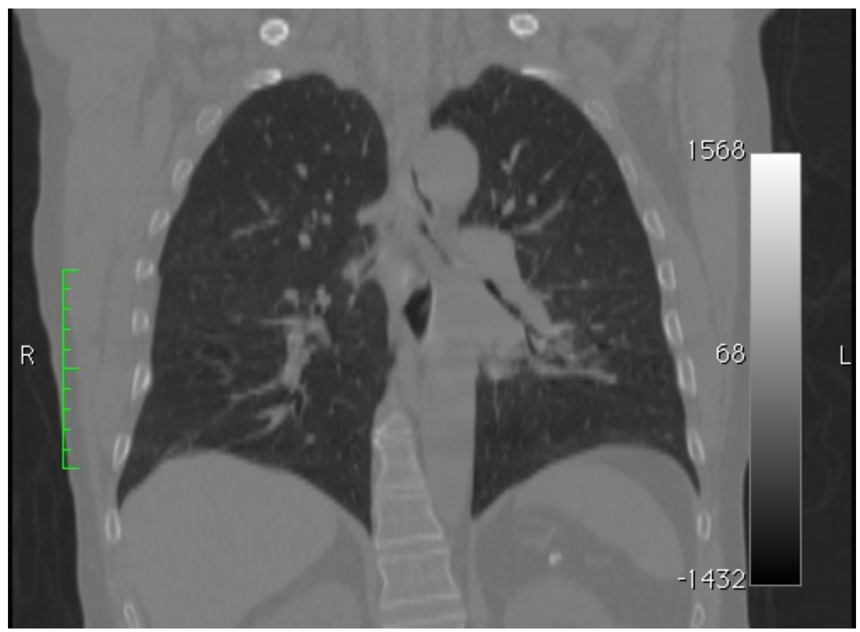
sagittal view

Comparing Projection and Tomographic Images

- Hounsfield's insight was that by imaging all the way around a patient we should have enough information to form a cross-sectional image
- *Sir* Godfrey Hounsfield shared the 1979 Nobel Prize with Allan Cormack (of FBP fame), funded by the EMI and the Beatles
- Radiographs typically have higher resolution but much lower contrast and no depth information (i.e. in CT section below we can see lung structure)

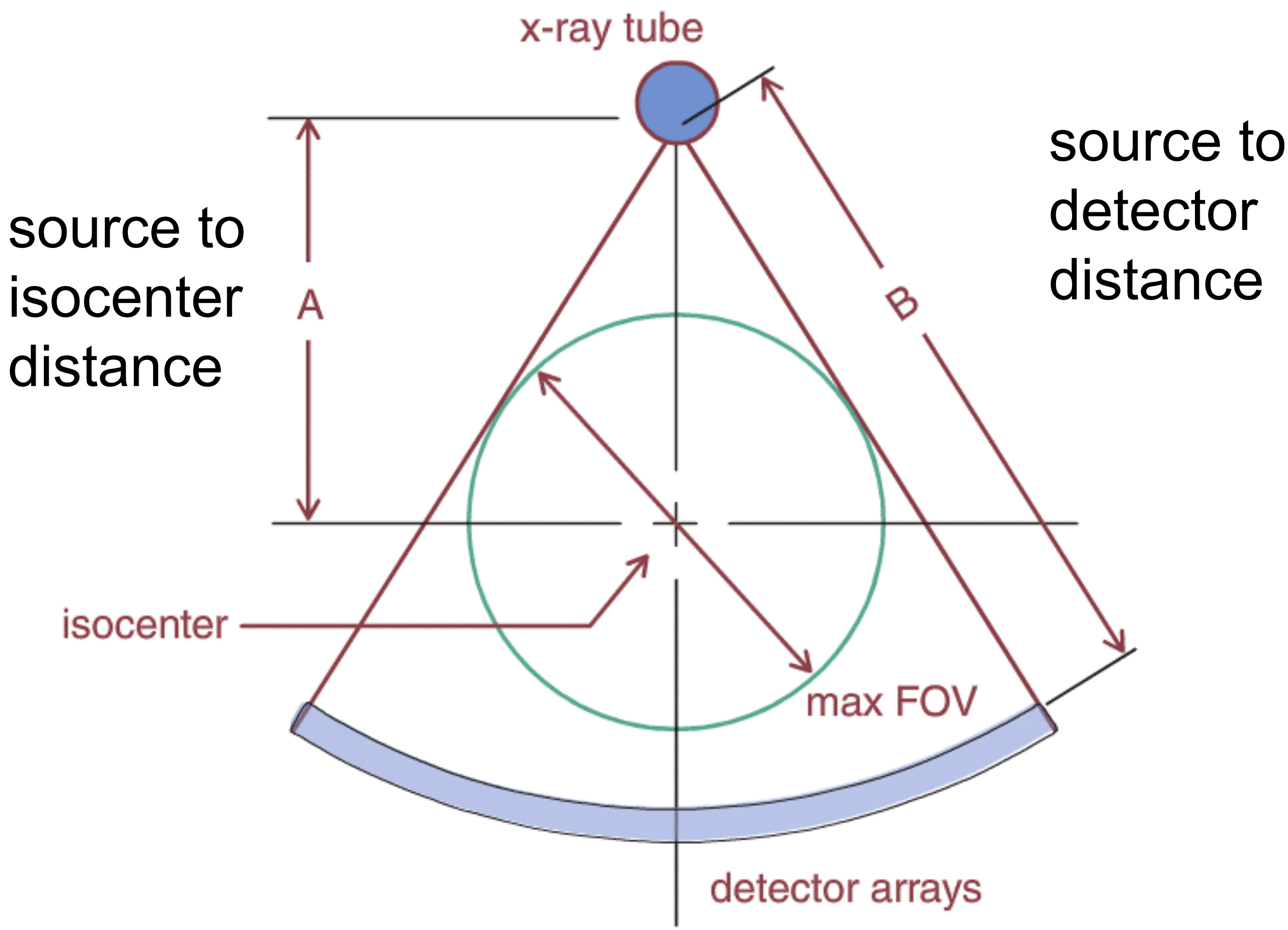


Chest radiograph

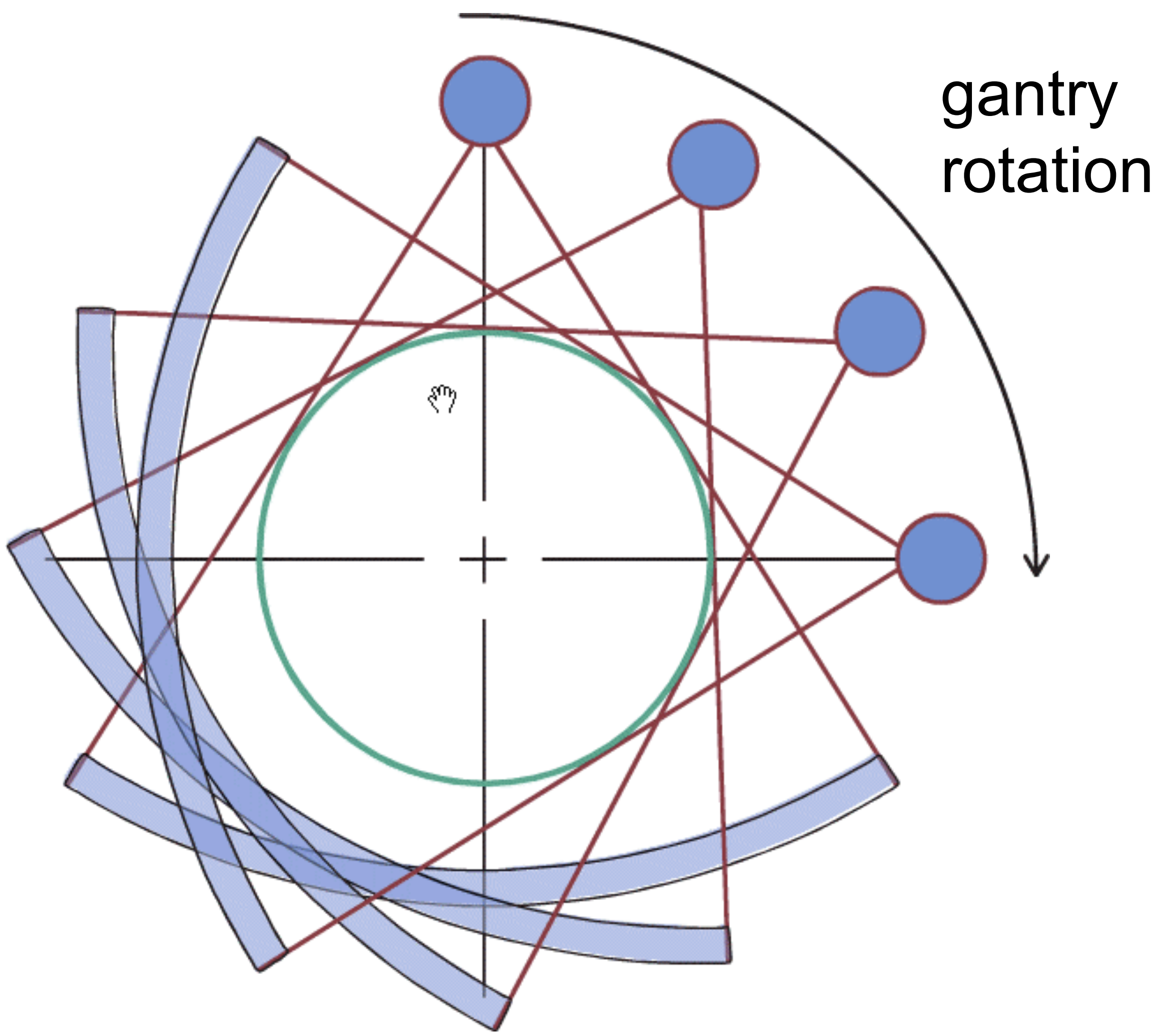


Coronal *section* of a 3D CT image volume

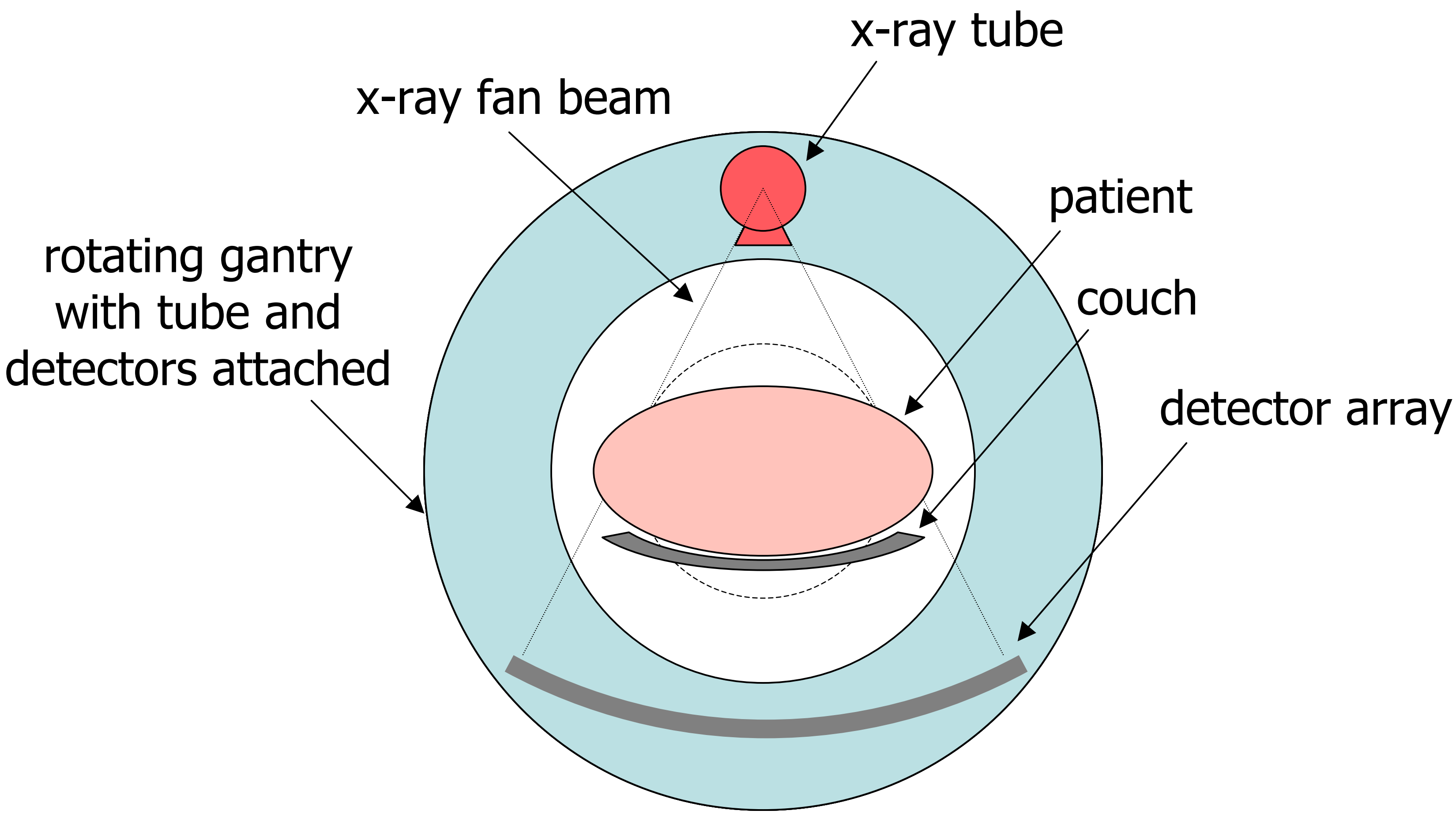
CT Scanner Geometry



CT Scanner Geometry

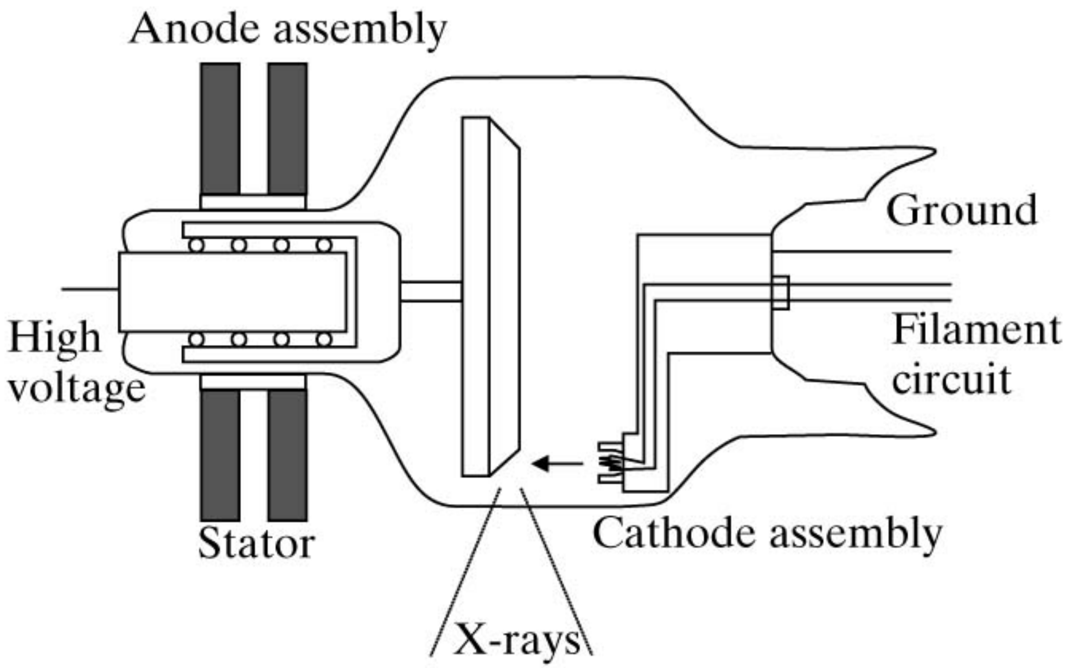


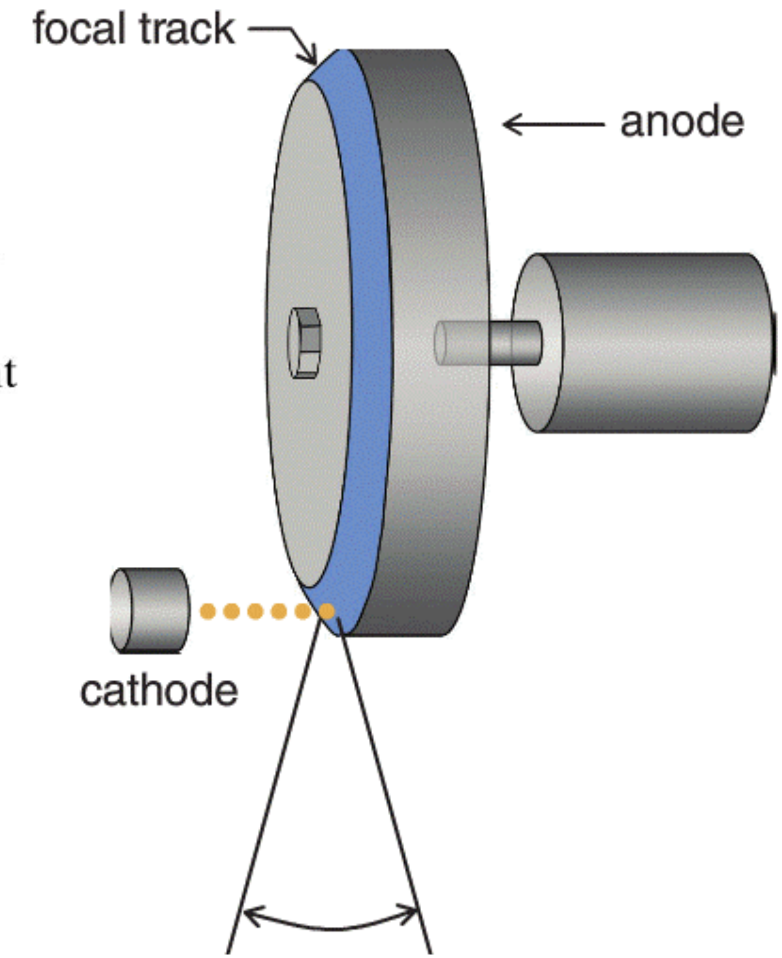
CT Scanner Components



- Data acquisition in CT involves making transmission measurements through the object at angles around the object.
- A typical scanner acquires 1,000 projections with a fan-beam angle of 30 to 60 degrees incident upon 500 to 1000 detectors and does this in <1 second.

CT X-ray Tube

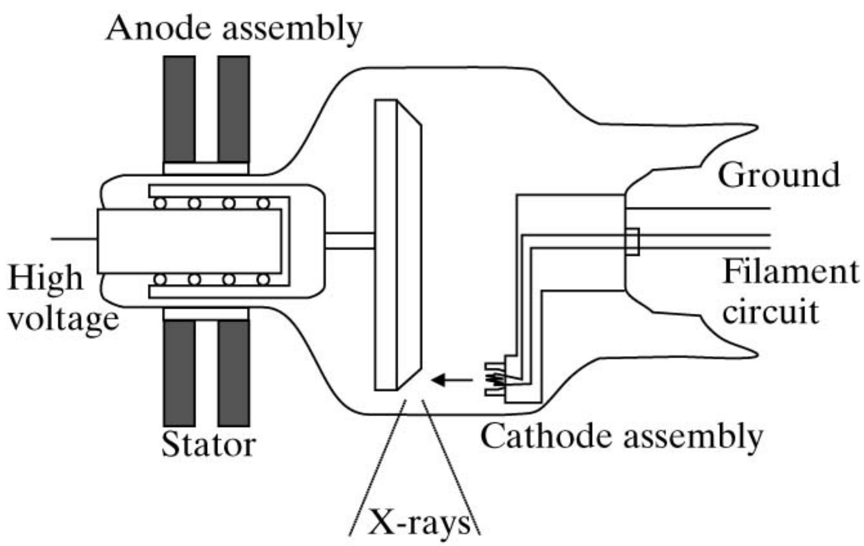


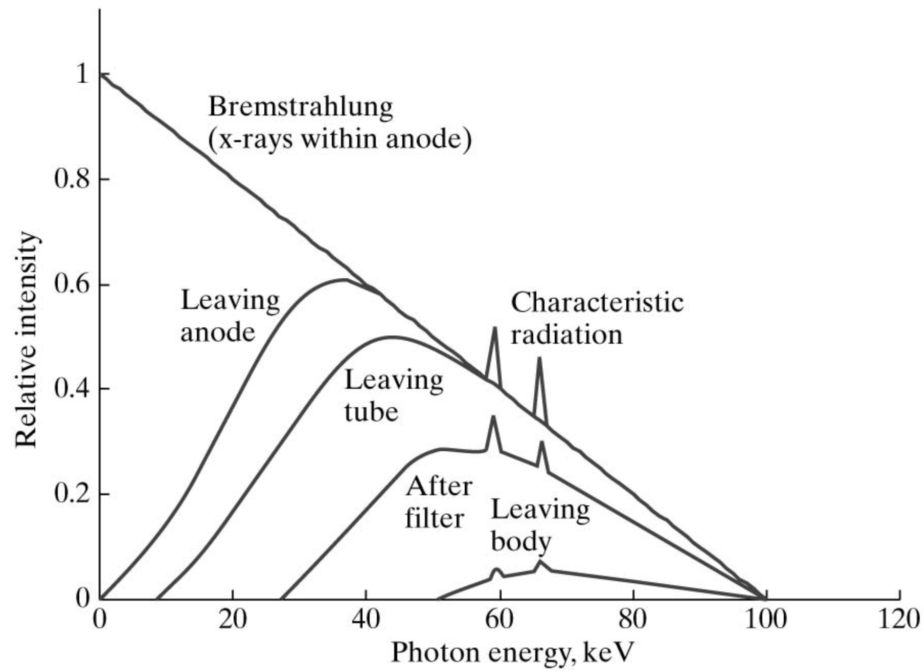


- In a vacuum assembly
- A resistive filament is used to 'boil off' electrons in the cathode with a carefully controlled current (10 to 500 mA)
- Free electrons are accelerated by the high voltage towards the anode

X-ray tubes

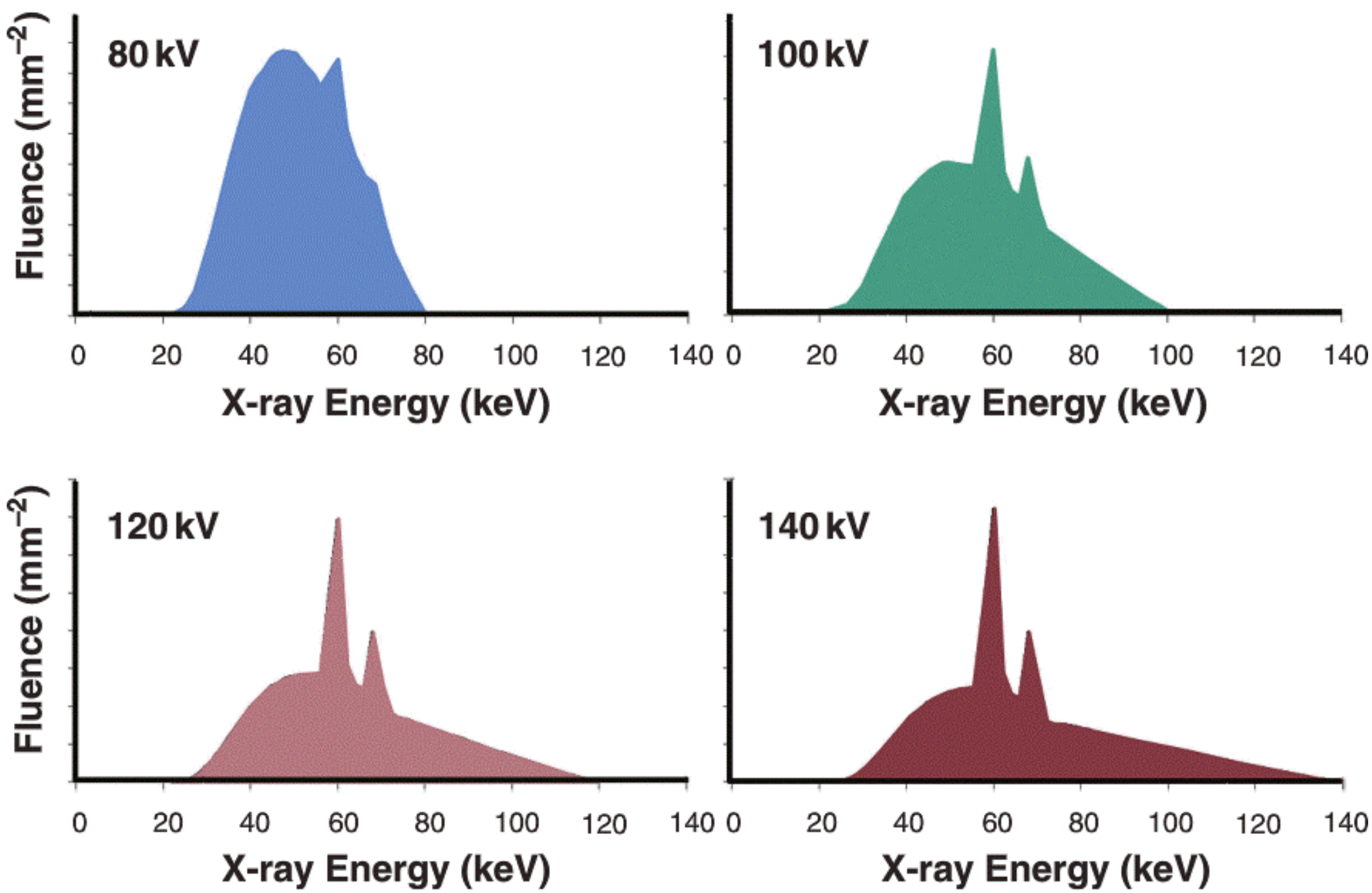
- Voltage determines maximum and x-ray energy, so is called the kVp (i.e. kilo-voltage potential), typically 90 kVp to 140 kVp for CT
- High-energy electrons smash into the anode
  - More than 99% energy goes into heat, so anode is rotated for cooling (3000+ RPM)
  - Bremmstrahlung then produces polyenergetic x-ray spectrum



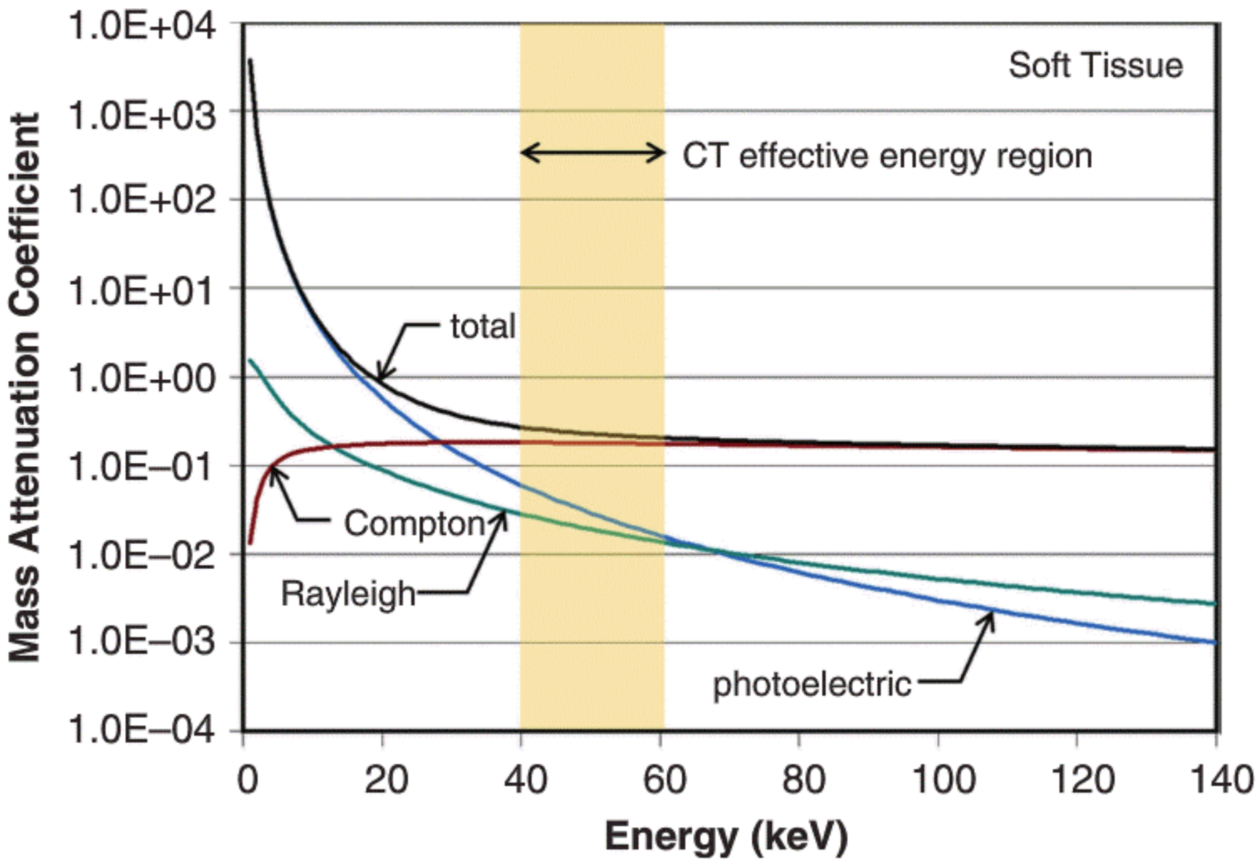


Typical X-ray spectra in CT

scaled to peak fluence

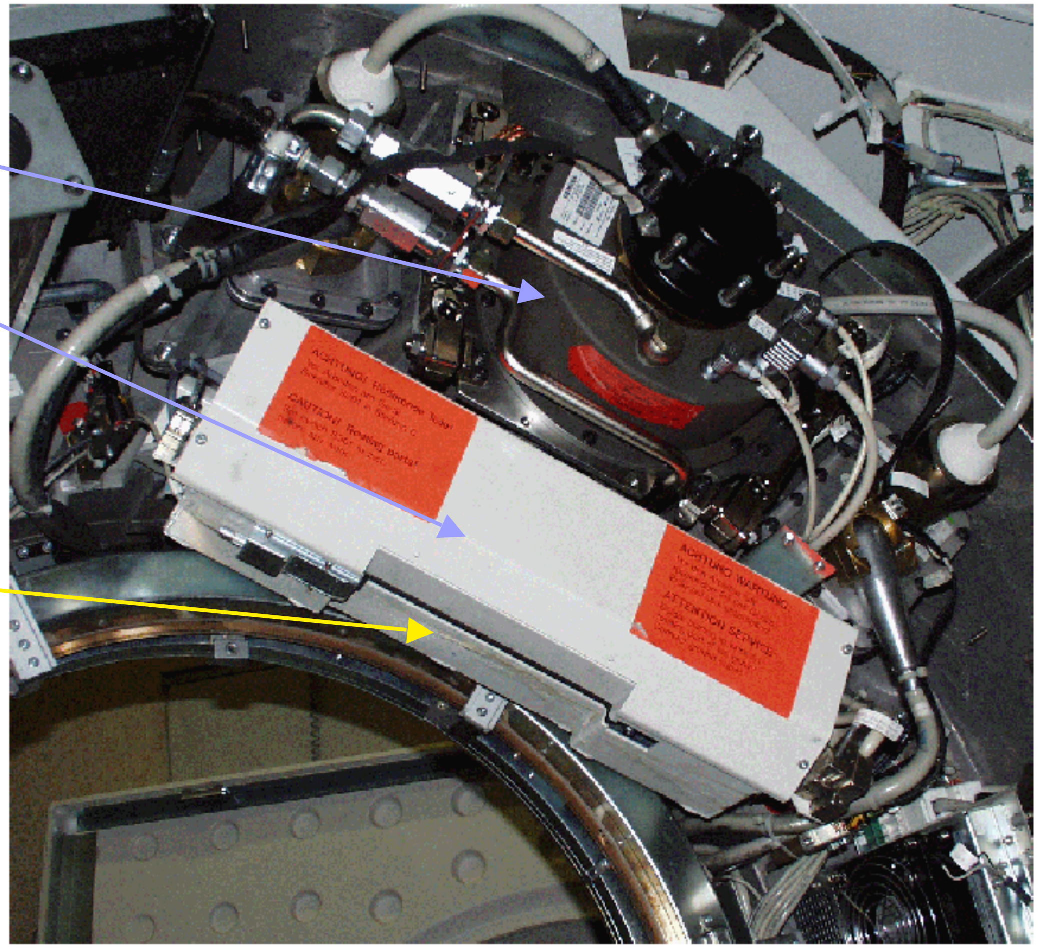


Mass attenuation coefficient versus energy

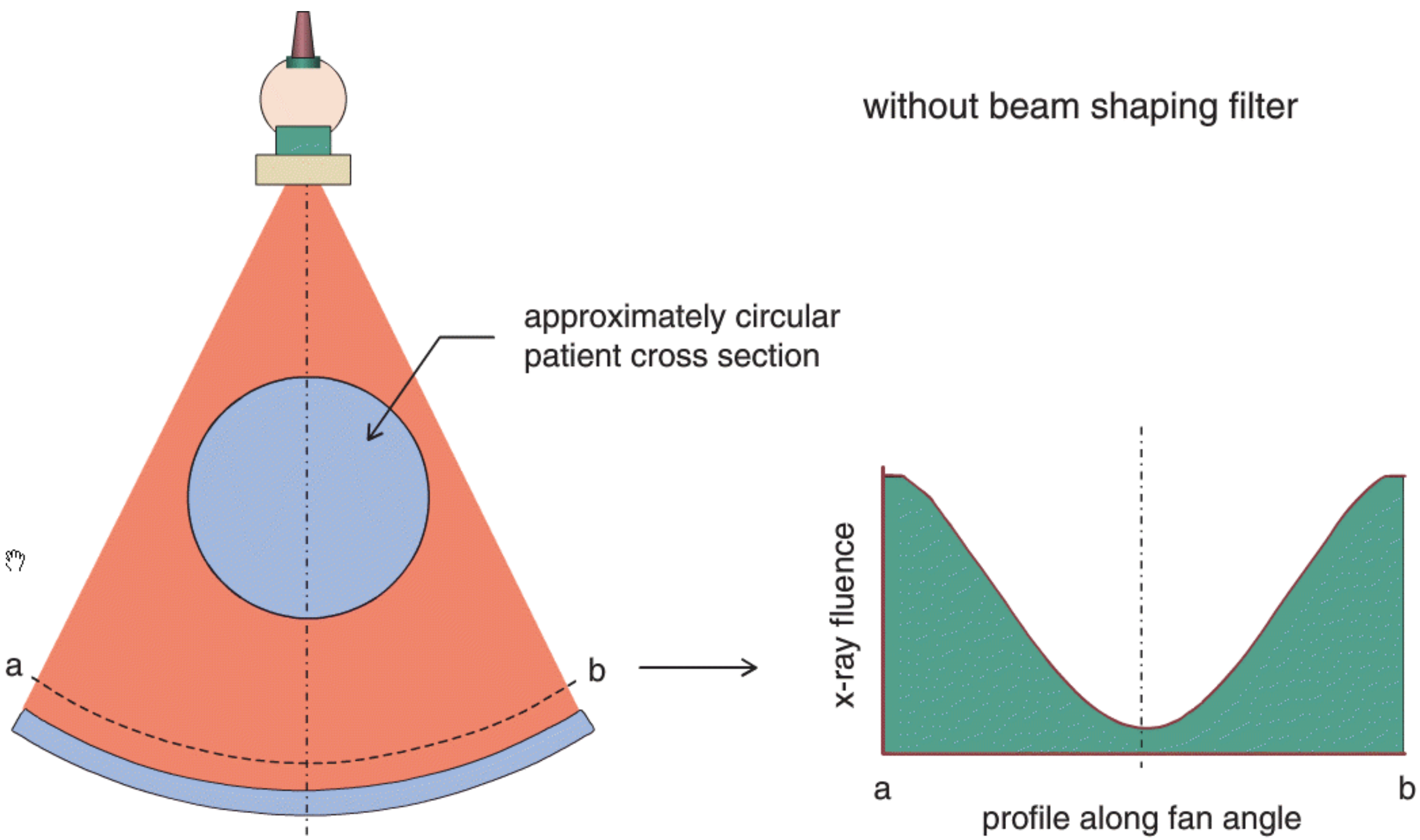


Pre-Patient Collimation

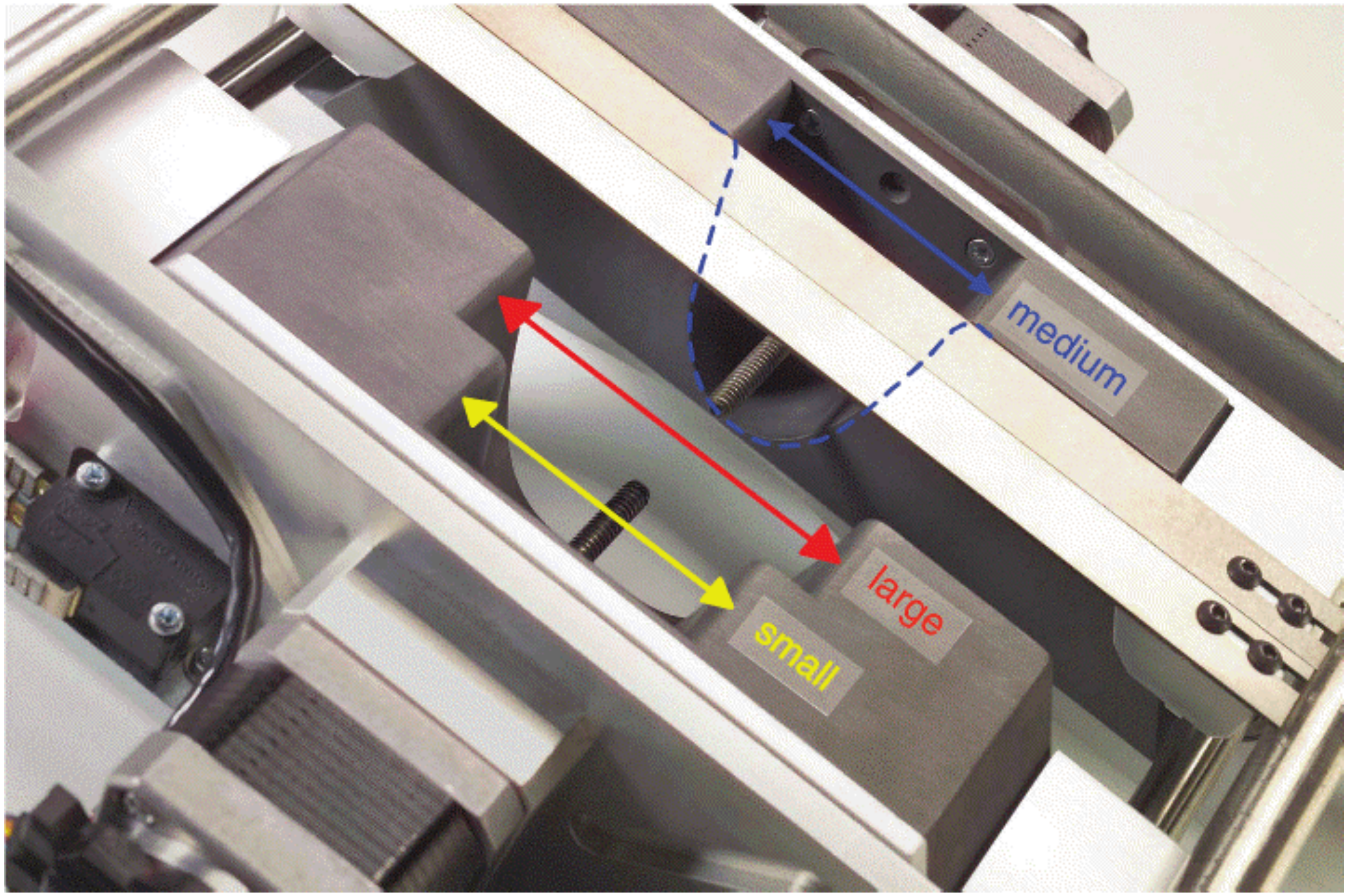
- Controls patient radiation exposure



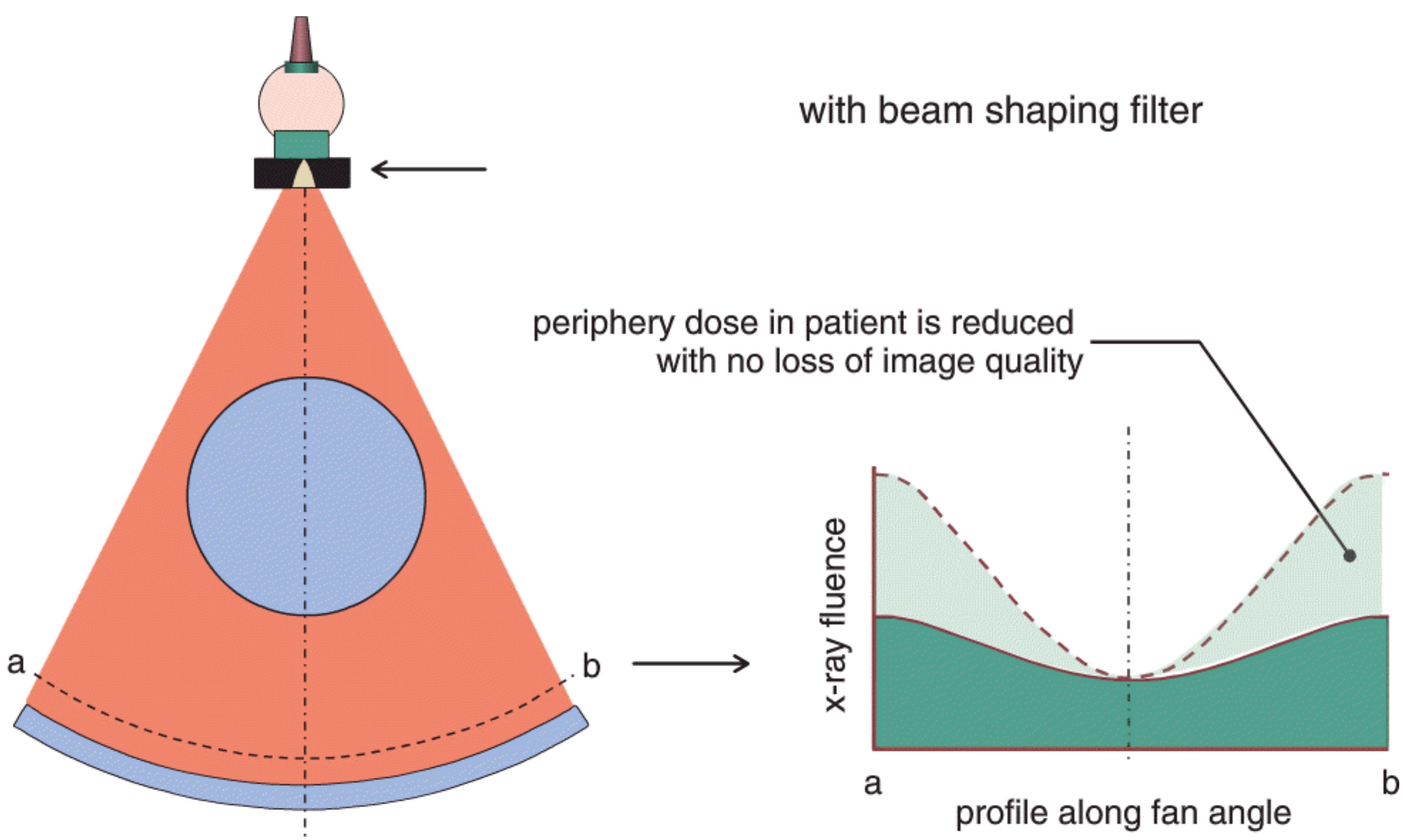
Need for x-ray beam shaping



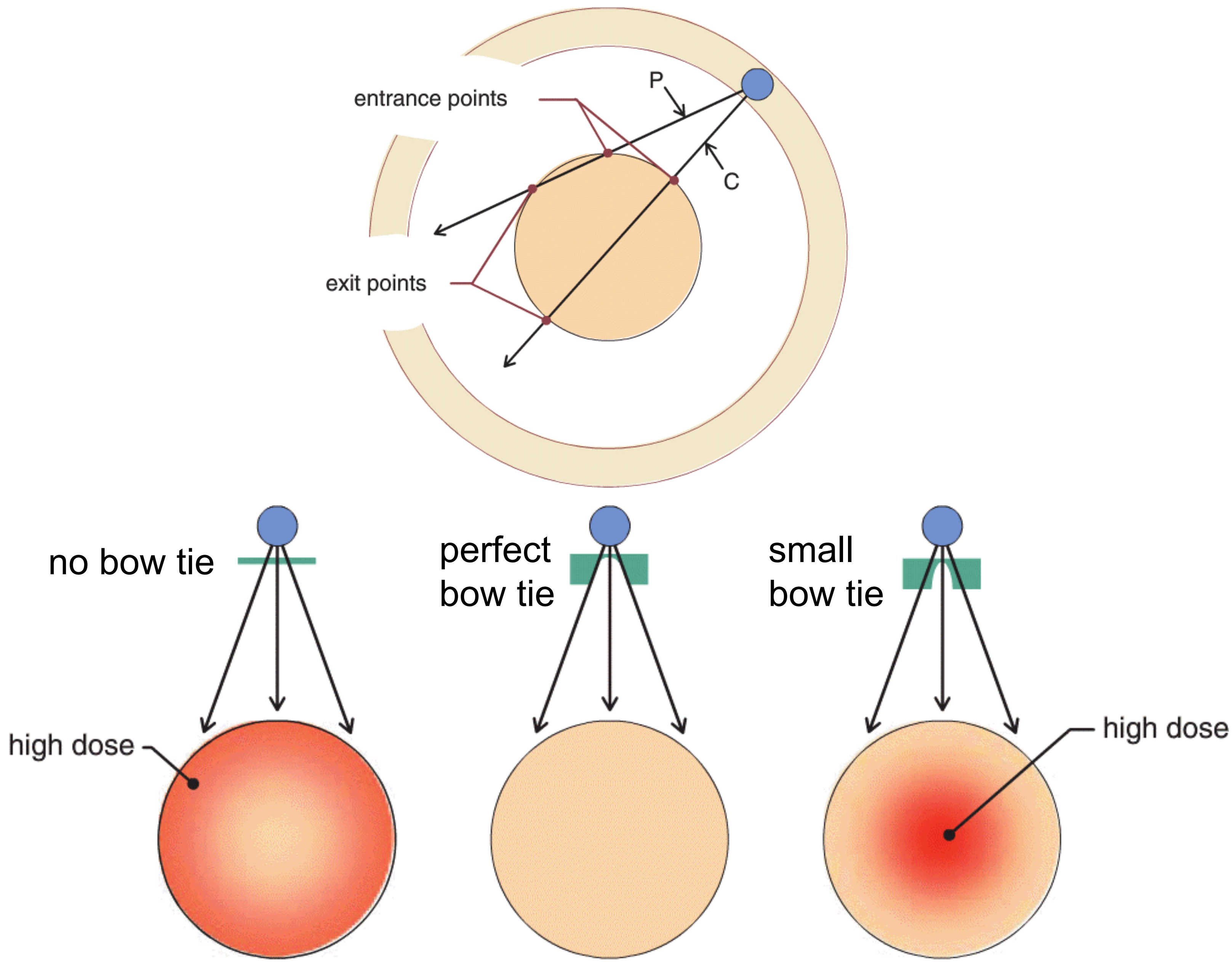
Addition of 'bow-tie' filters for beam shaping



Use of 'Bow-tie' beam shaping

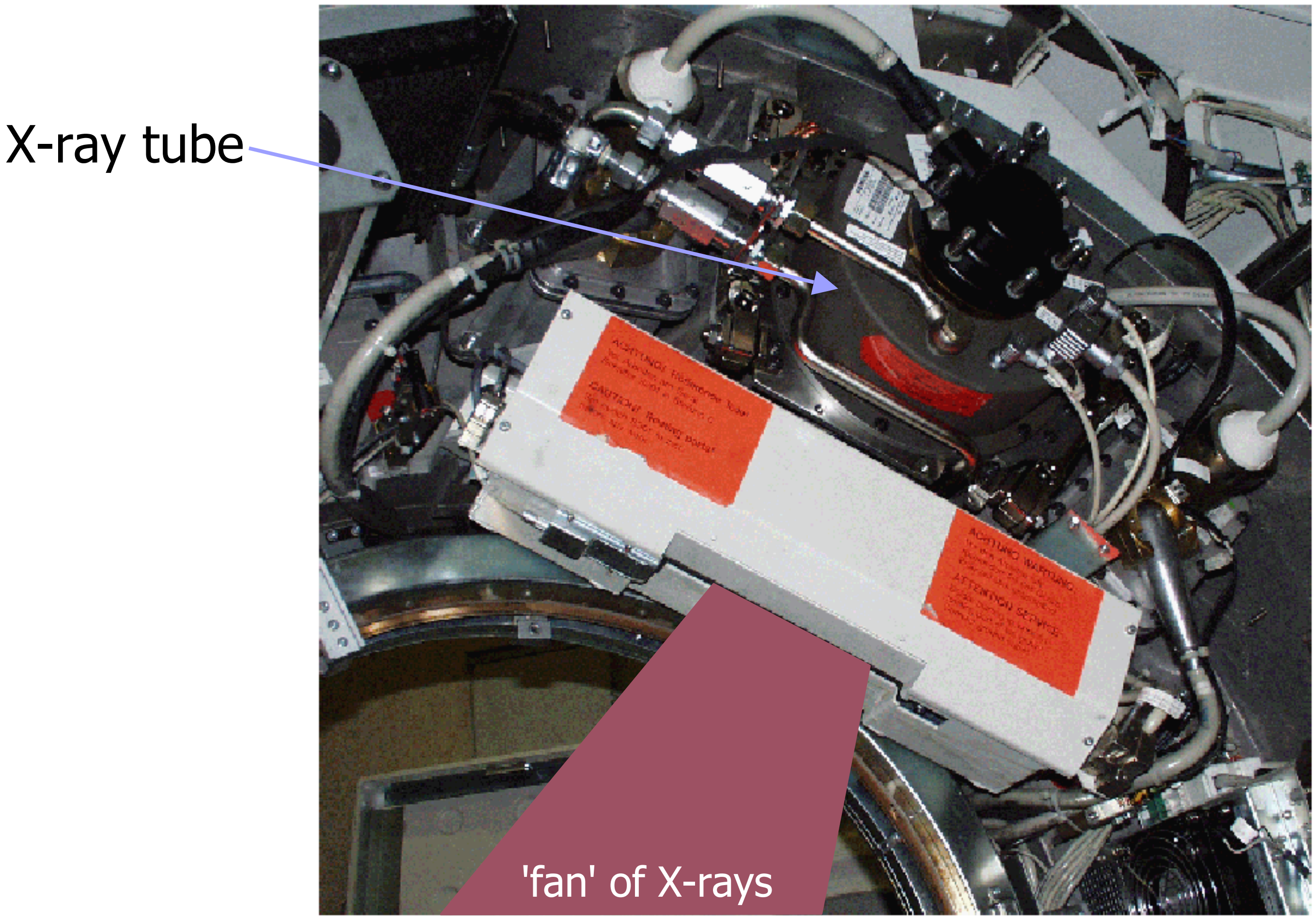


Radiation dose considerations



Pre-Patient Collimation

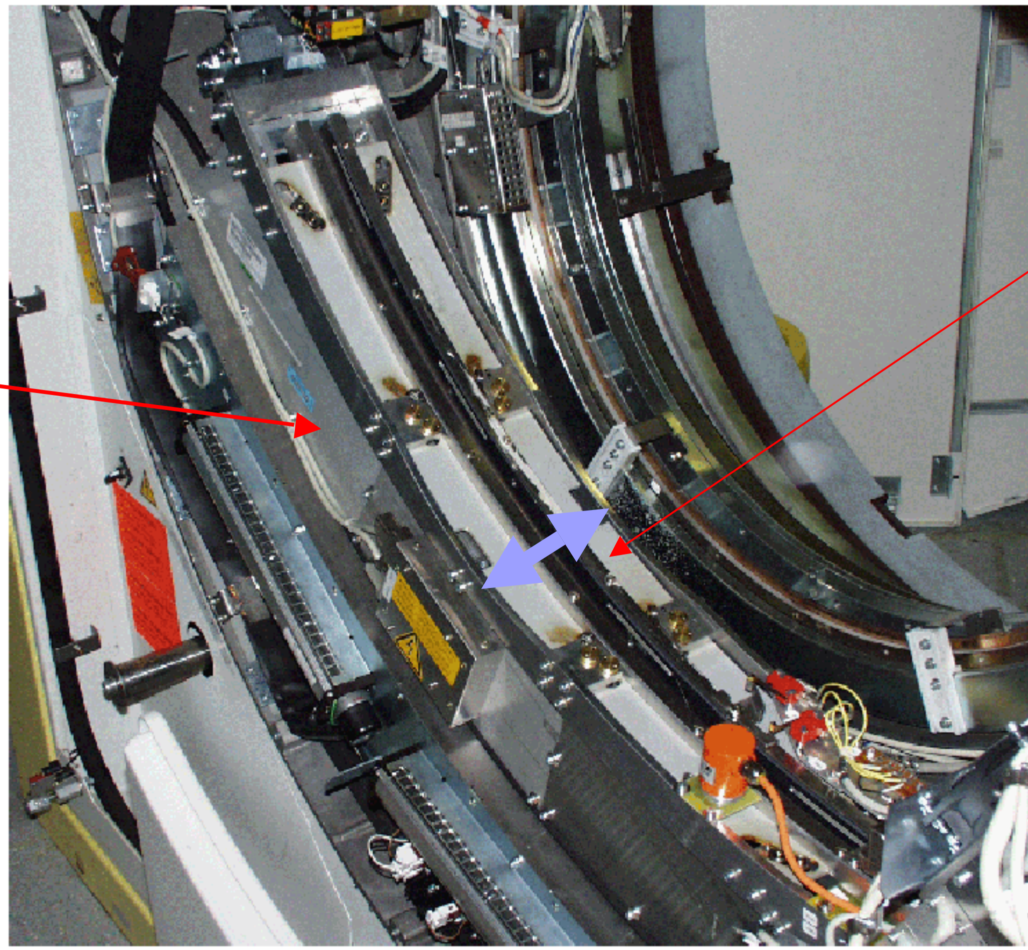
- Controls patient radiation exposure



X-ray Detector Assembly

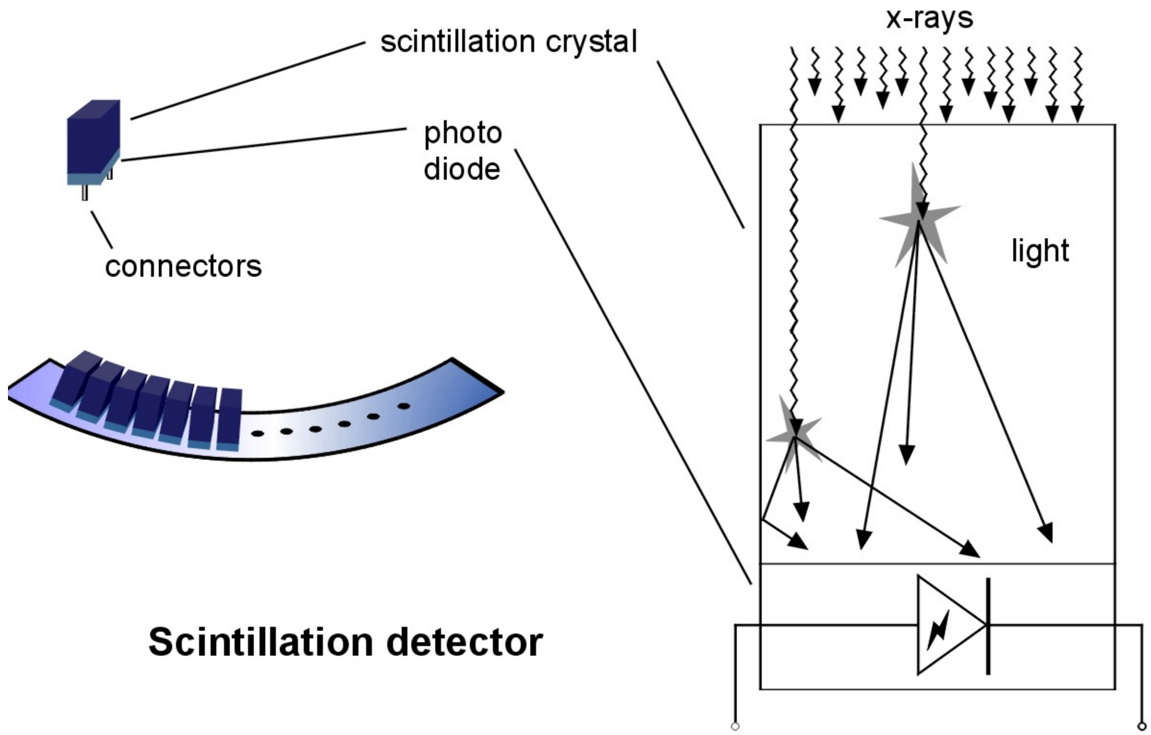
collimators

detectors



X-ray CT Detectors

- The detectors are similar to those used in digital flat-panel imaging systems: scintillation followed by light collection
- The scintillator converts the high-energy photon to a light pulse, which is detected by photo diodes

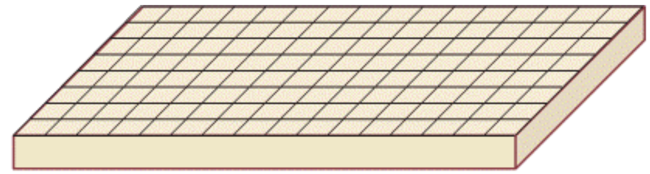


Scintillation detector

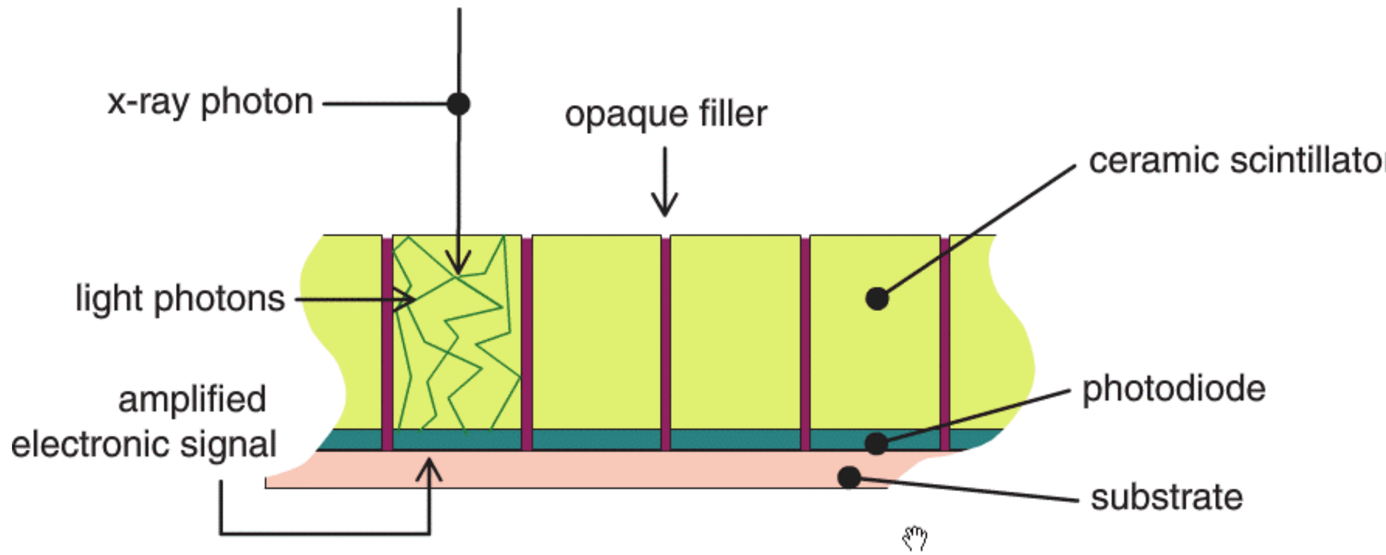
X-ray CT Detectors

Typically composed of rare-earth crystals (e.g. Gd_2O_2S)

Sintered to increase density



perspective view

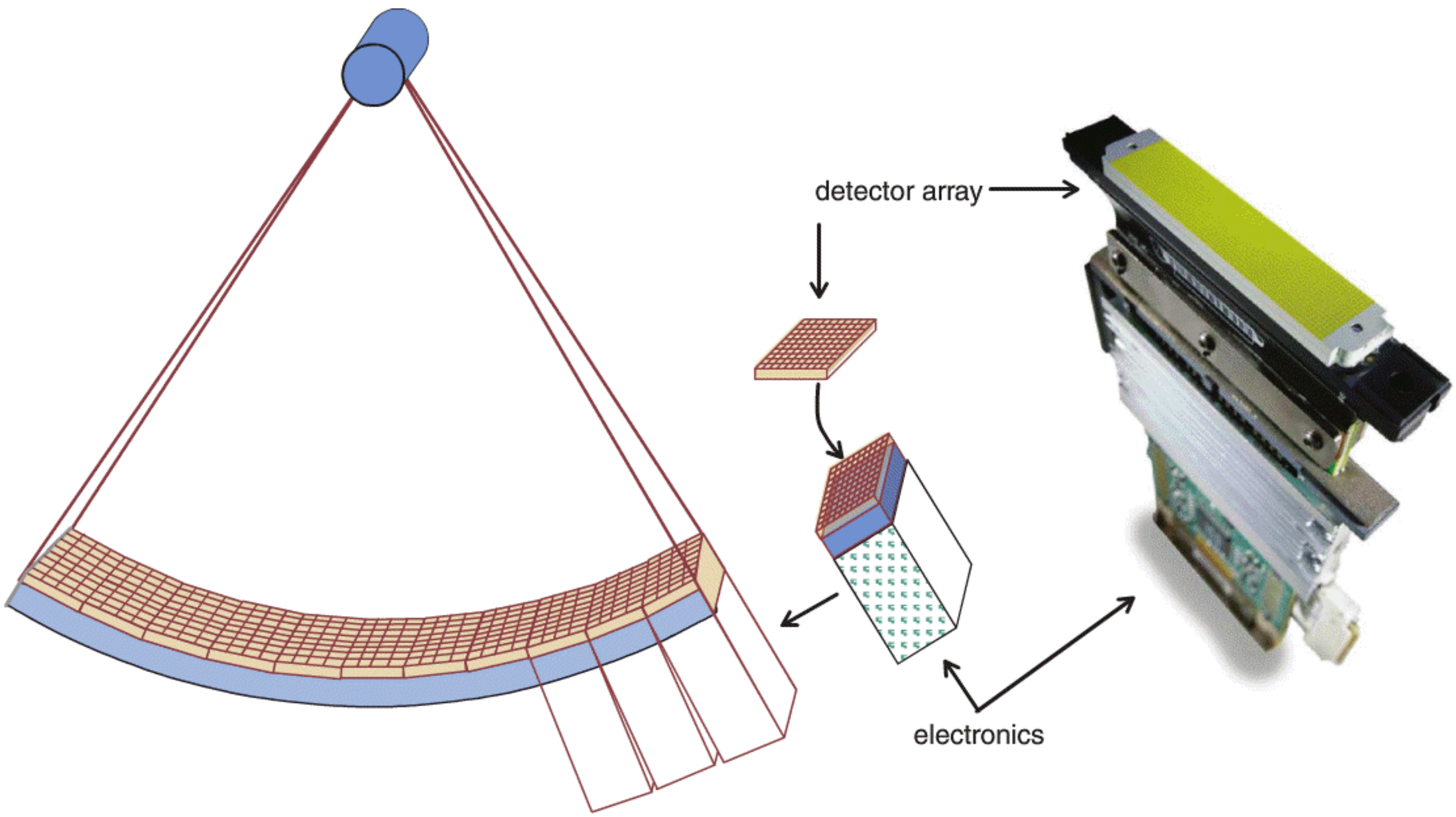


side view

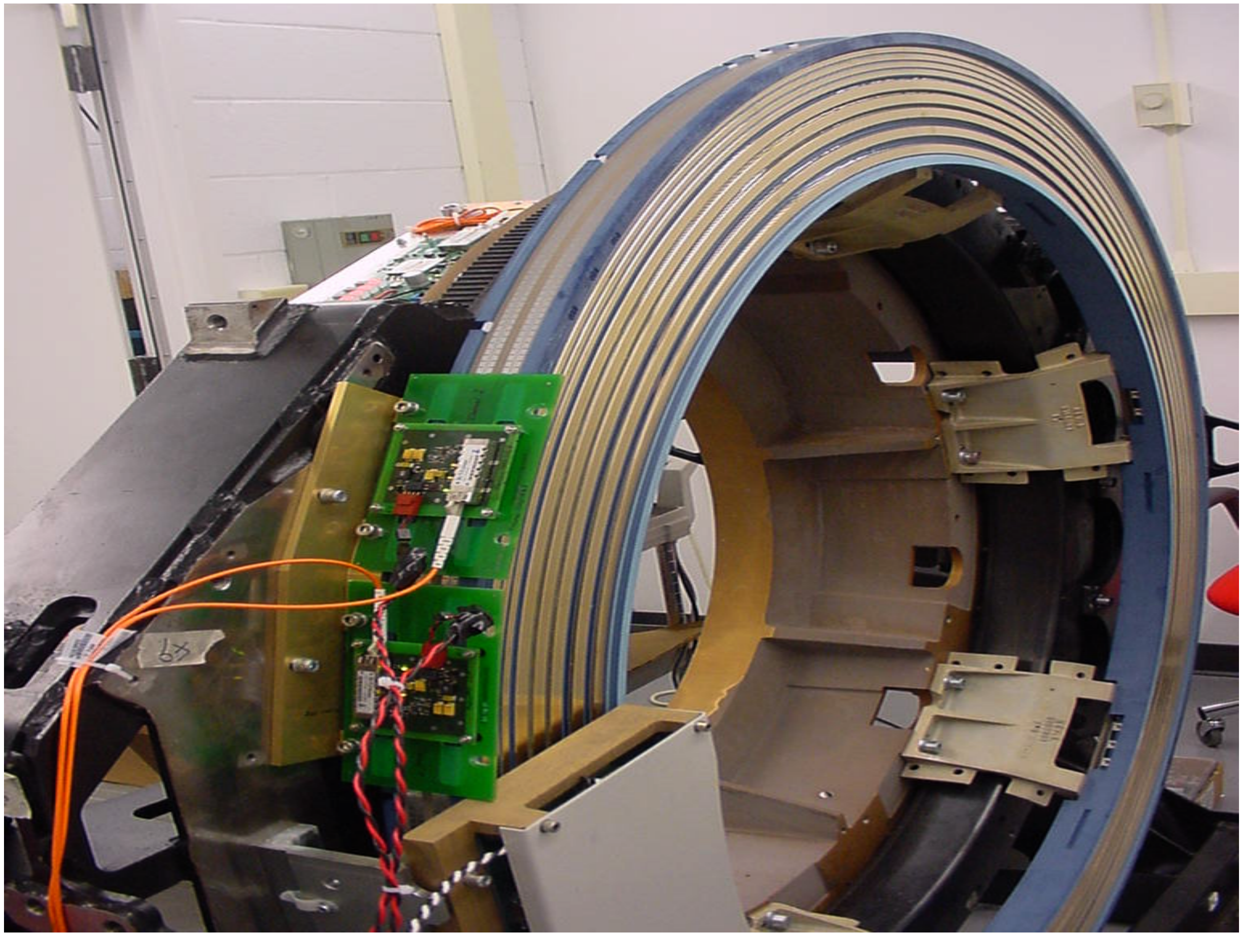
X-ray CT Detectors

Detector module sits on a stack of electronic modules

- pre-amp
- ADC
- voltage supply



Gantry Slip Rings



- Allows for continuous rotation

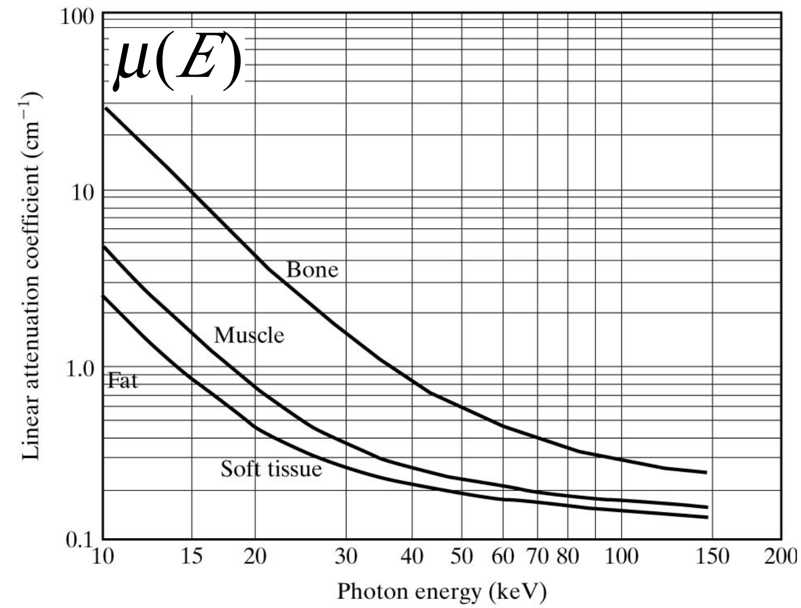
CT Scanner in Operation

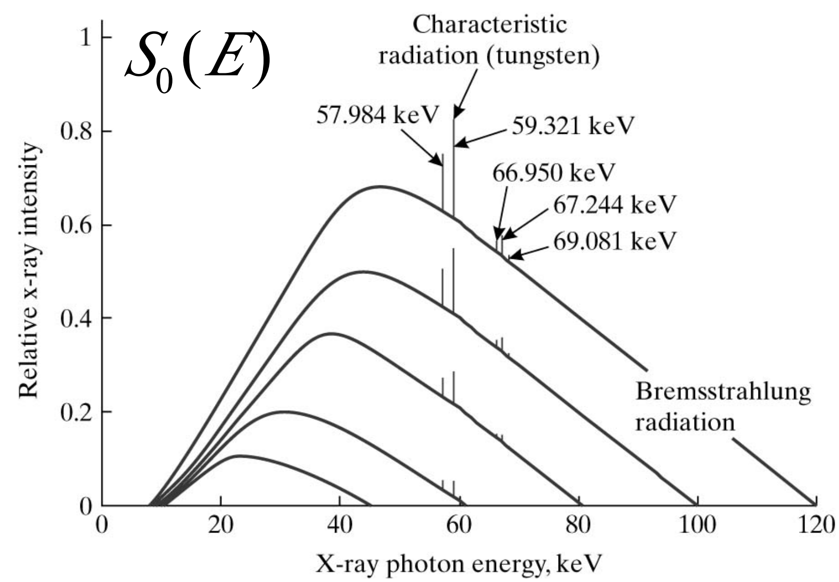
- 64-slice CT, weight ~ 1 ton, speed 0.33 sec (180 rpm)

Narrow-beam Polyenergetic Attenuation

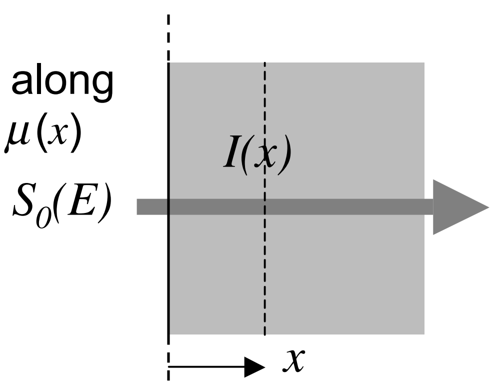
- The attenuation depends on material (thus position of material) and energy
- With bremsstrahlung radiation, there is a weighted distribution of energies
- We combine previous results to get the imaging equation

$$I(x) = \int_{E=0}^{E_{\max}} E' S_0(E') e^{-\int_0^x \mu(x', E')\, dx'} dE'$$





beam intensity along a line with $\mu = \mu(x)$



Imaging Equation

- Similar to x-ray projection systems (ignoring geometric effects etc.) for intensity at a detector location d

$$I_d = \int_0^{E_{\max}} S_0(E) E e^{-\int_0^d \mu(s,E)\,ds} \, dE$$

- In this case I_d is our measured data, and we want to recover an image of $\mu(x,y)$
- Unfortunately, the integration over energy presents a mathematically intractable inverse problem
- We work around this approximately by assuming an *effective energy*

$$\bar{E} = \frac{\int_0^{E_{\max}} E S(E)\,dE}{\int_0^{E_{\max}} S(E)\,dE}$$

Approximate Imaging Equation

- Using an effective energy, we can write the imaging equation as

$$I_d = I_0 e^{-\int_0^d \mu(s,\bar{E})ds}$$

- A further simplification comes from defining $g_d \triangleq -\ln\left(\frac{I_d}{I_0}\right)$

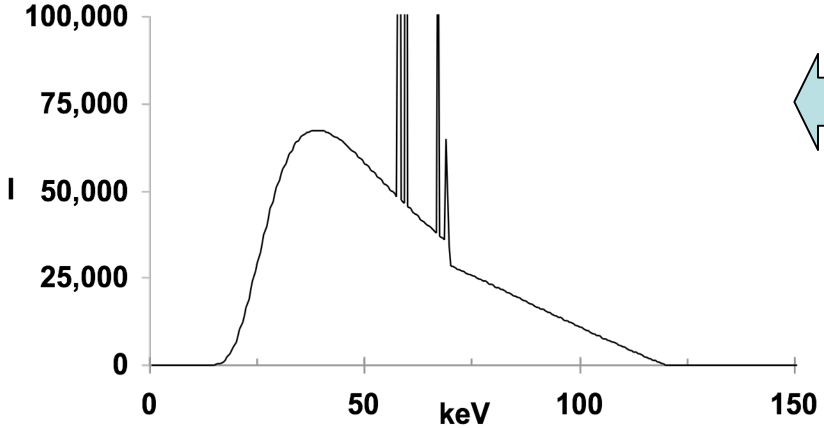
- Giving an x-ray transform $g_d = -\int_0^d \mu(s,\bar{E})ds$

(we can solve *this* imaging equation)

  – We need to measure the reference intensity I_0, typically done with a detector at the edge of the fan
  – Although we can use FBP, the effective energy will be object dependent, so the reconstructed $\mu(x,y)$ will only be approximate

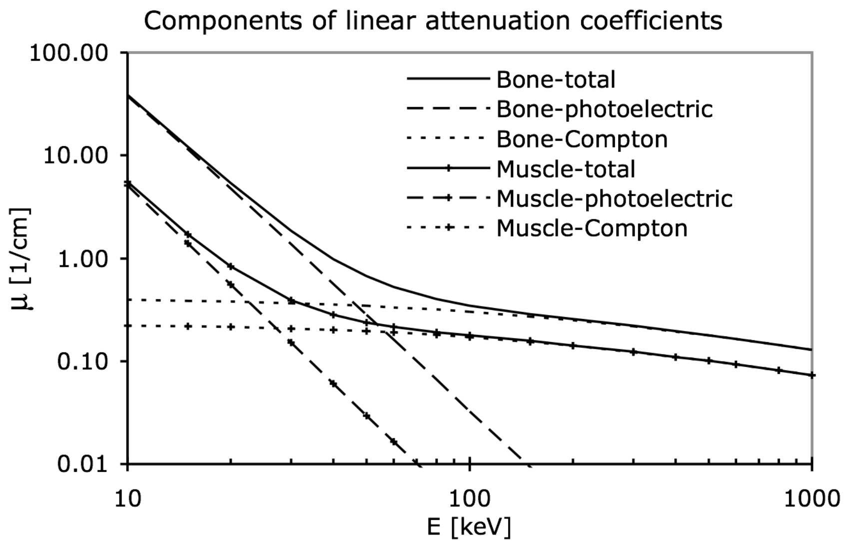
X-ray CT Image Values

- With CT attempt to determine μ(x,y), but due to the bremsstrahlung spectrum we have a complicated weighting of μ (x,y) at different energies, which will change with scanner and patient thickness due to differential absorption.



Input x-ray bremsstrahlung spectrum (intensity vs. photon energy) for a commercial x-ray CT tube set to 120 kVp

Energy dependent linear attenuation coefficients (μ(x,y)) for bone and muscle



CT Numbers or Hounsfield Units

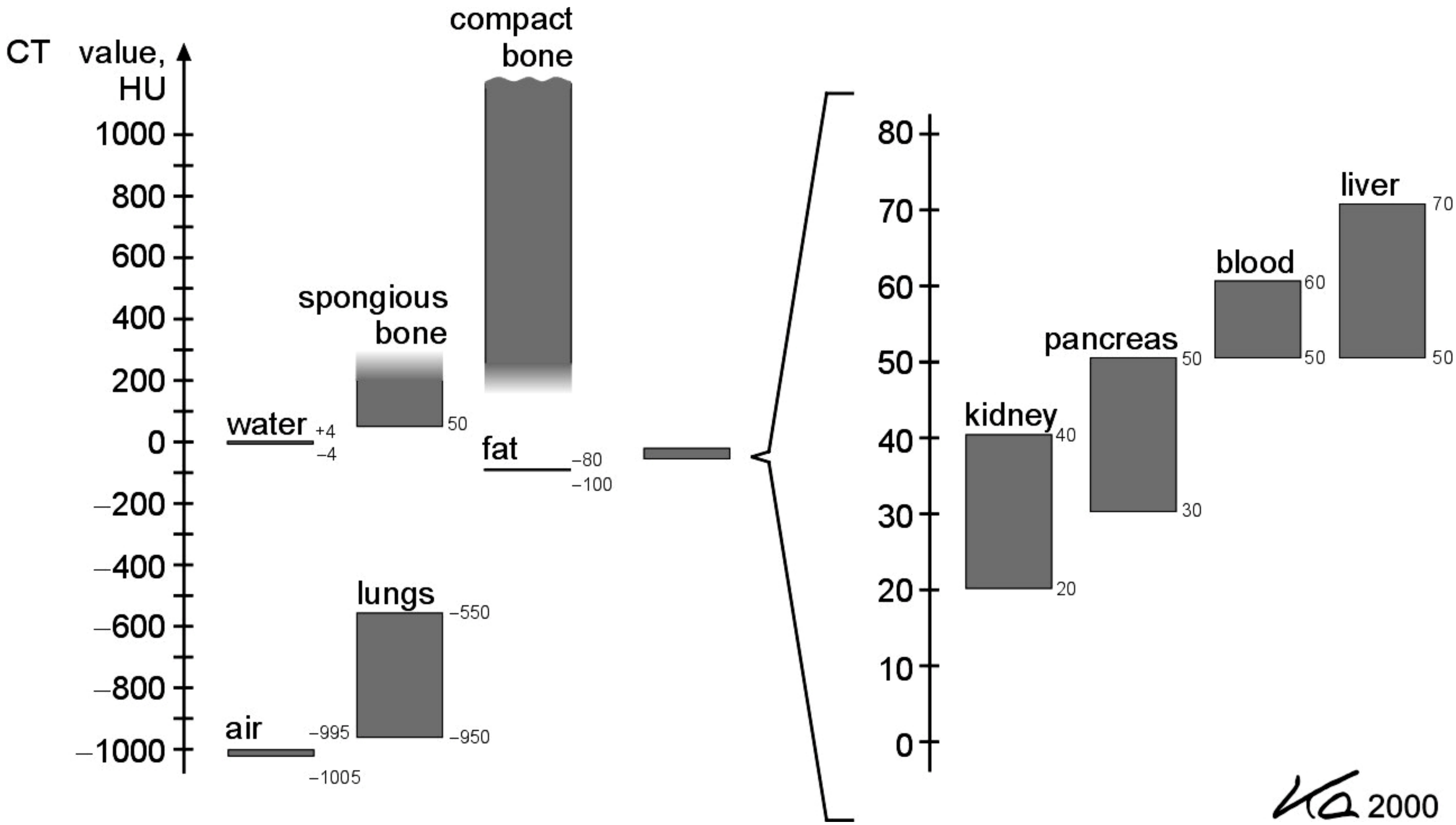
- We can't solve the real inverse problem since we have a mix of densities of materials, each with different Compton and photoelectric attenuation factors at different energies, and a weighted energy spectrum
- The best we can do is to use an *ad hoc* image scaling
- The CT number for each pixel, (x,y) of the image is scaled to give us a fixed value for water (0) and air (-1000) according to:

$$\mathrm{CT}(x,y) = 1000\left[\frac{\mu(x,y)-\mu_{water}}{\mu_{water}}\right]$$

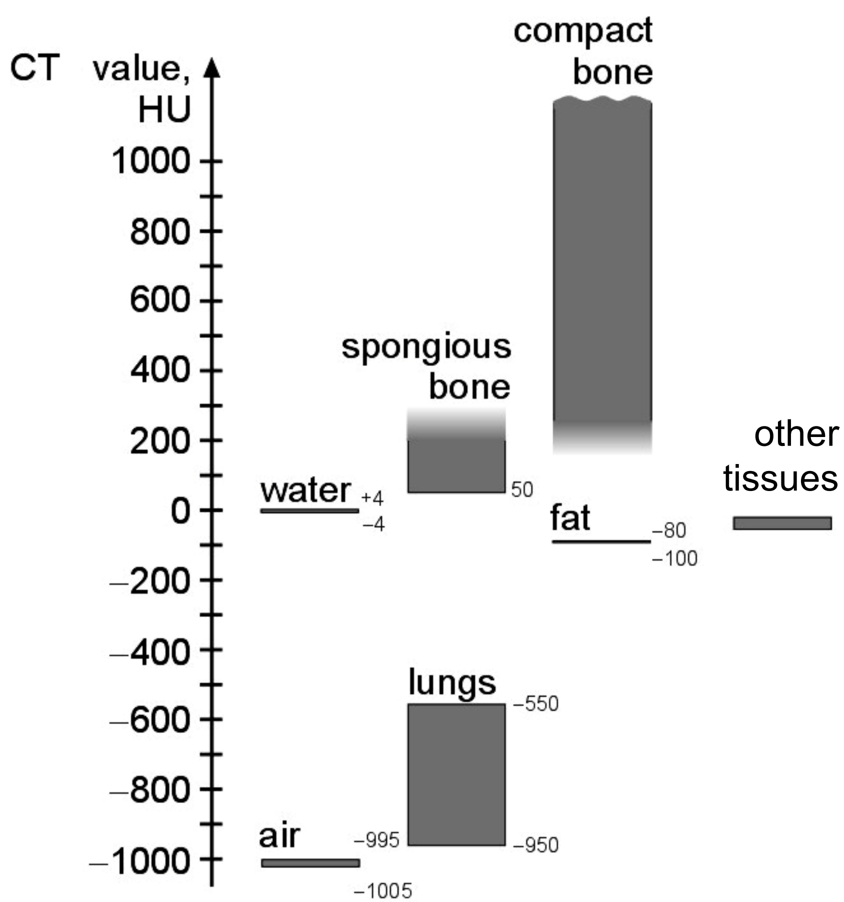
- $\mu(x, y)$ is the reconstructed attenuation coefficient for the voxel, μ_{water} is the attenuation coefficient of water and CT(x,y) is the CT number (using *Hounsfield units*) of the voxel values in the CT image

CT Numbers

- Typical values in Hounsfield Units

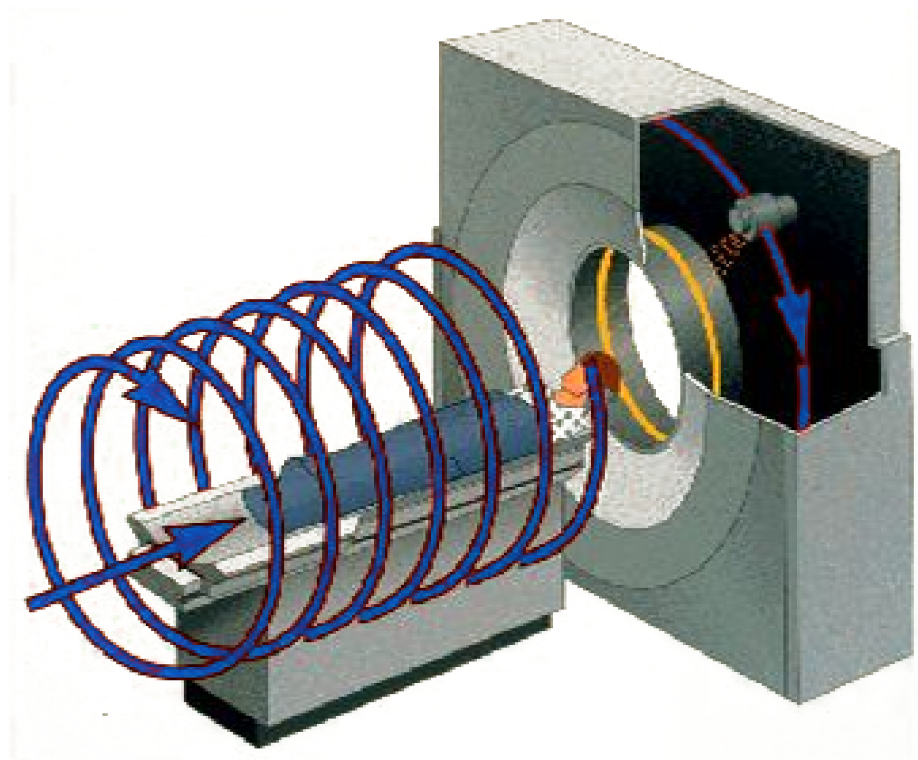


CT scan showing 'apparent' density



Helical CT Scanning

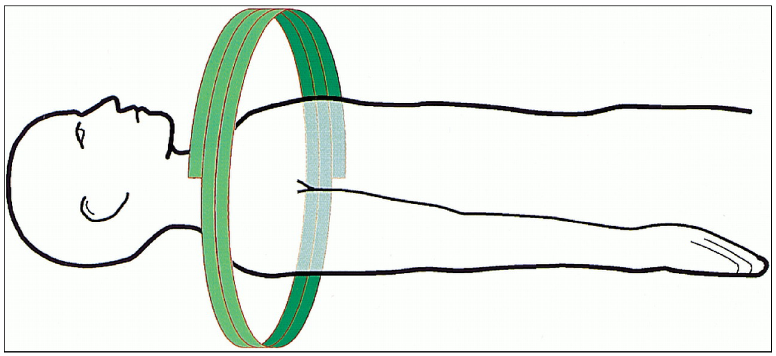
- The patient is transported continuously through gantry while data are acquired continuously during several 360-deg rotations
- The ability to rapidly cover a large volume in a single-breath hold eliminates respiratory misregistration and reduces the volume of intravenous contrast required



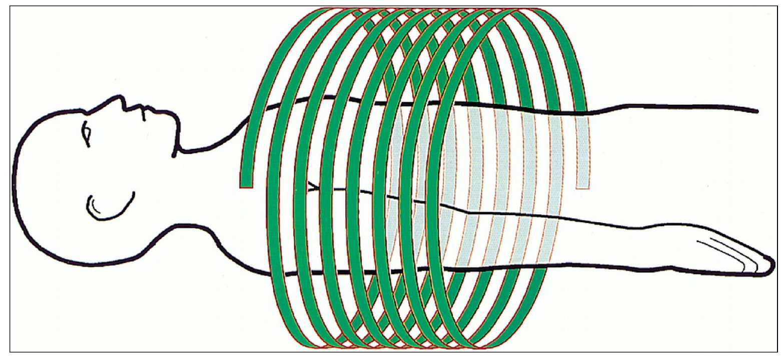
Pitch

$$\text{pitch} = \frac{\text{table travel per rotation}}{(\text{number detectors}) \text{ x } (\text{detector width})} = \frac{\text{table travel per rotation}}{\text{acquisition beam width}}$$

slingle slice example



Pitch = 1

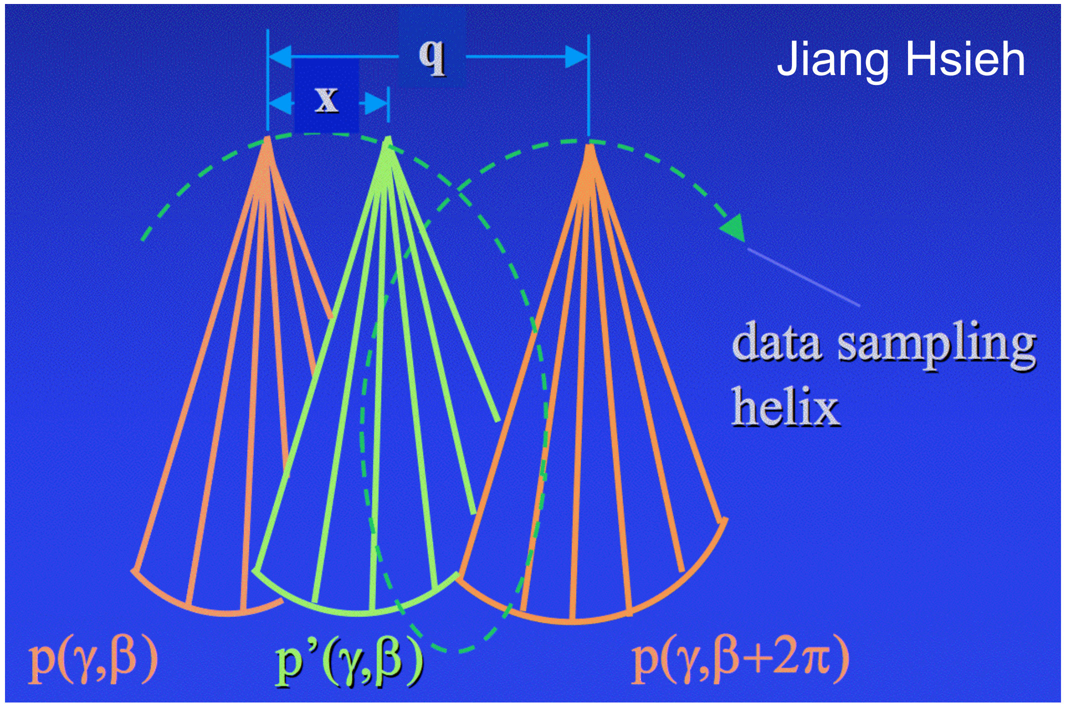


Pitch = 2

- A pitch of 1.0 is roughly equivalent to axial (i.e. one slice at a time) scanning
  - best image quality in helical CT scanning
- A pitch of less than 1.0 involves overscanning
  - some slight improvement in image quality, but higher radiation dose to the patient
- A pitch greater than 1.0 is not sampling enough, relative to detector axial extent, to avoid artifacts
  - Faster scan time, however, often more than compensates for undersampling artifacts (i.e. patient can hold breath so no breathing artifacts).

Image Reconstruction from Helical data

- Samples for the plane-of-reconstruction are estimated using two projections that are 2π apart

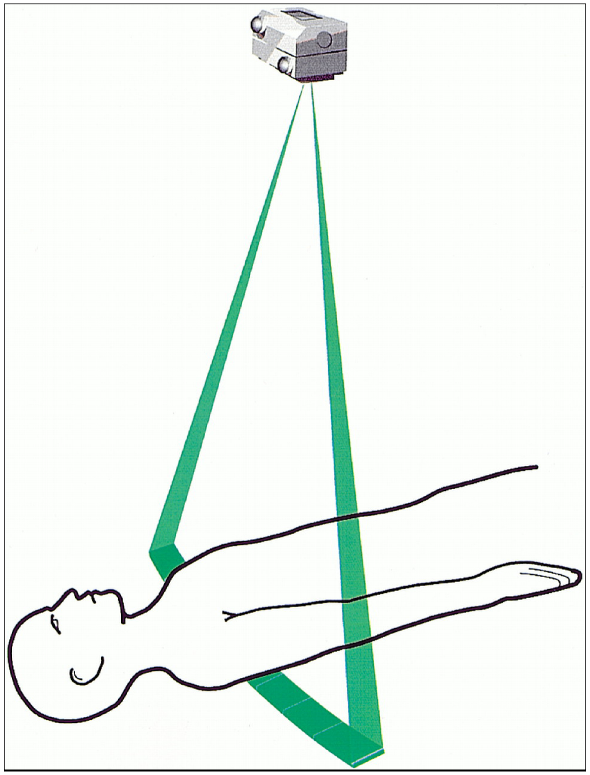


$$p'(\gamma,\beta) = wp(\gamma,\beta) + (1-w)\,p(\gamma,\beta+2\pi)$$

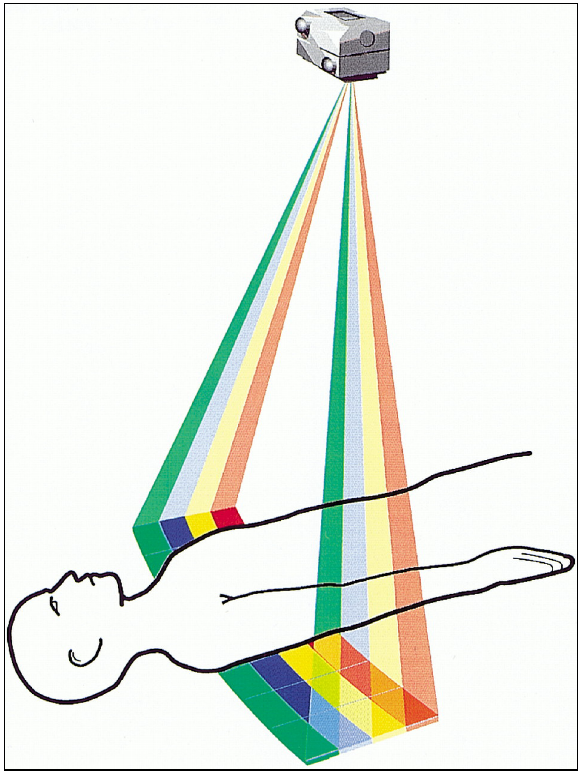
where $w = (q-x)/q$

Single versus Multi-row Detectors

- Can image multiple planes at once



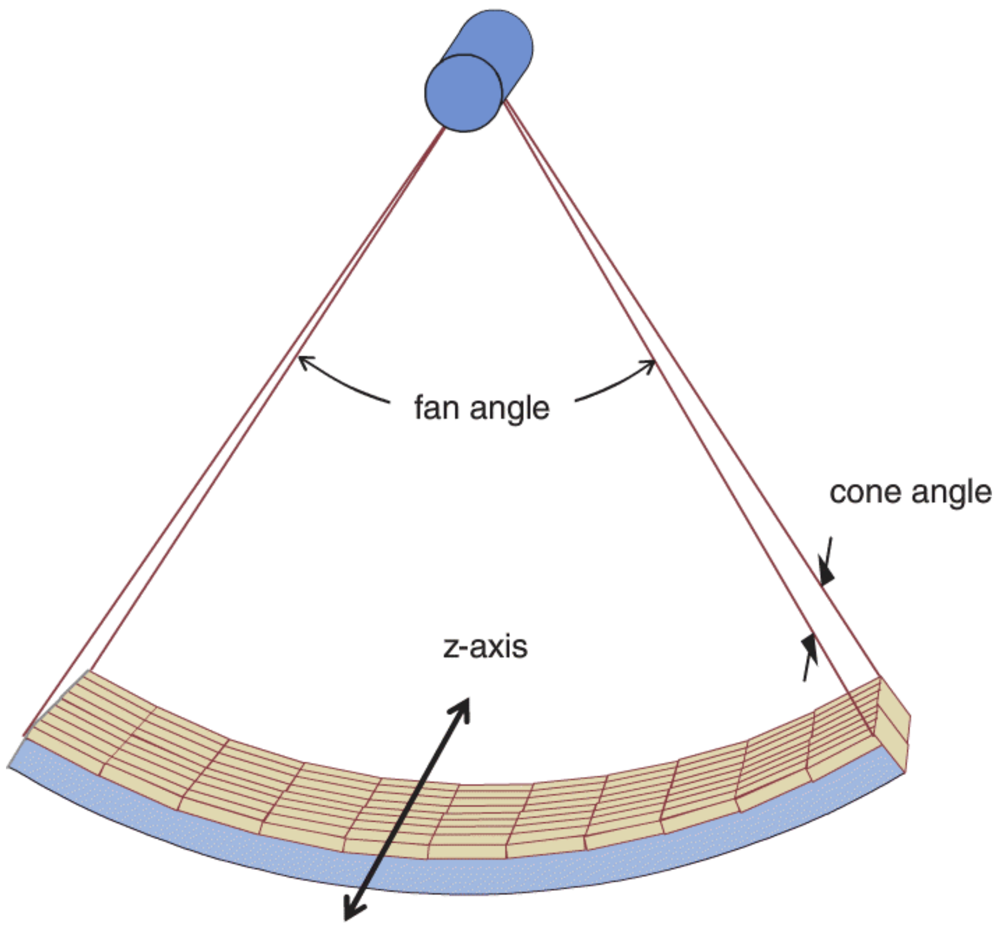
1 detector row



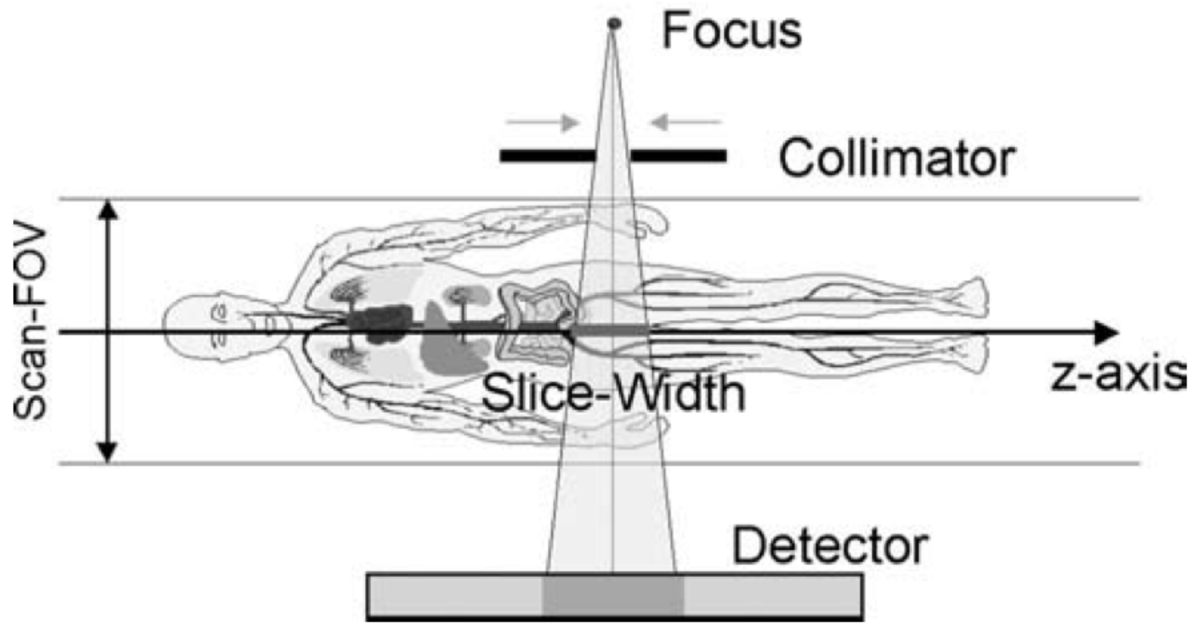
4 detector rows

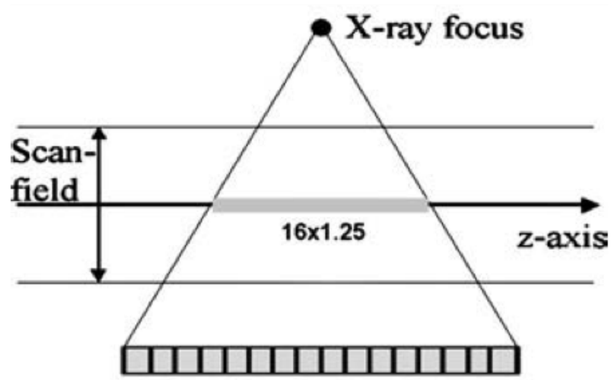
Single versus Multi-row Detectors

- Can image multiple planes at once



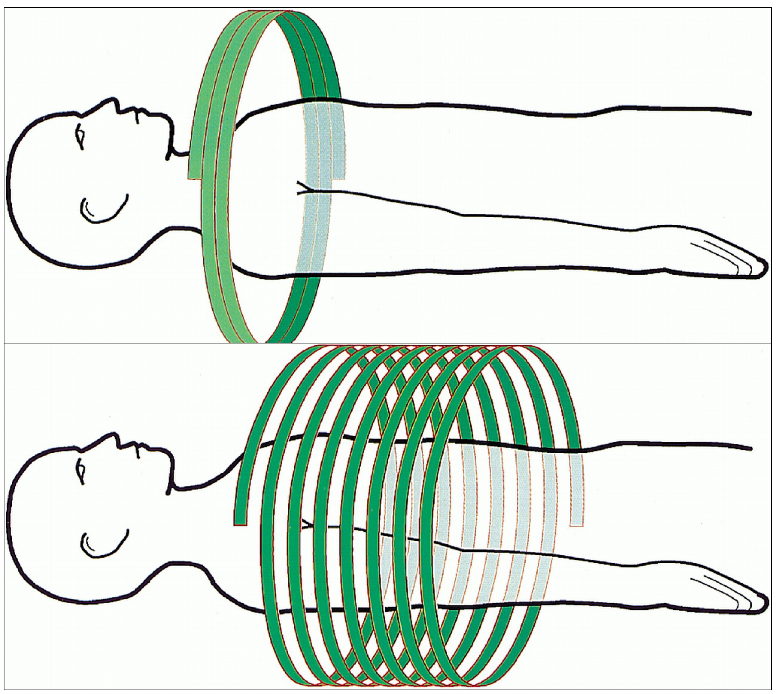
Multi-row Detectors



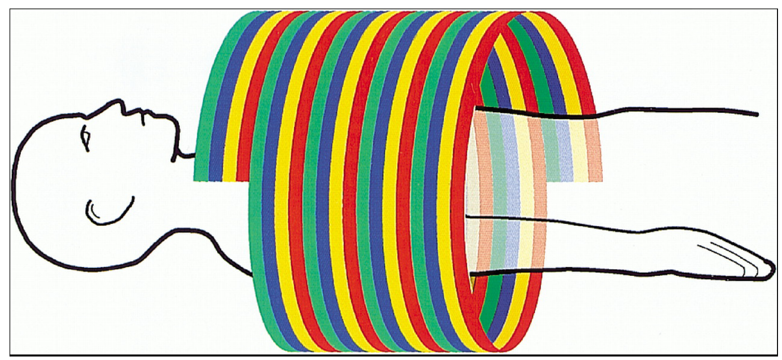


Helical Multi-Detector CT (MDCT)

- Fastest possible acquisition mode -- same region of body scanned in fewer rotations, even less motion effects
- Single row scanners have to either scan longer, or have bigger gaps in coverage, or accept less patient coverage
- The real advantage is reduction in scan time



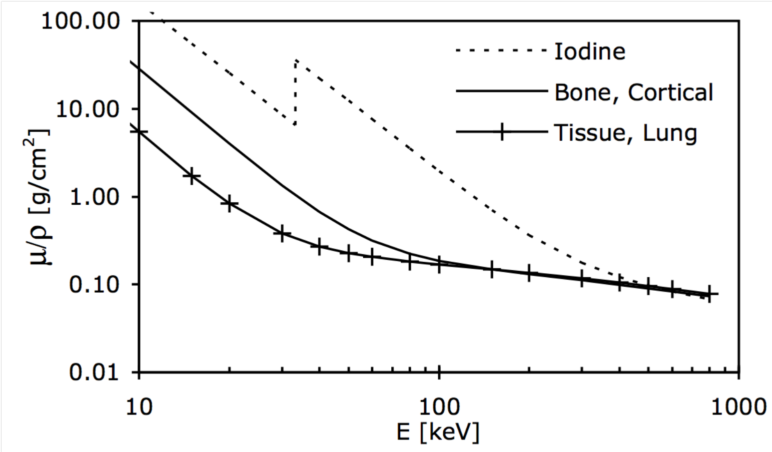
1 detector row: pitch 1 and 2



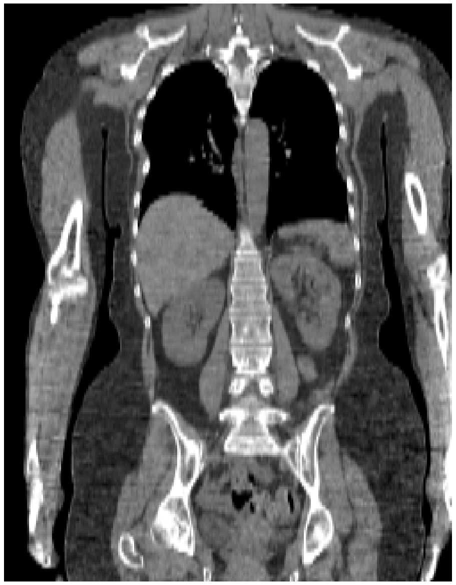
4 detector rows: pitch 1

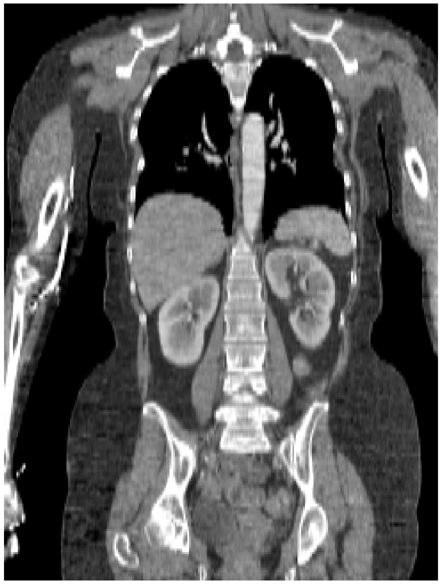
Contrast Agents

- Iodine- and barium-based contrast agents (very high Z) can be used to enhance small blood vessels and to show breakdowns in the vasculature
- Enhances contrast mechanisms in CT
- Typically iodine is injected for blood flow and barium swallowed for GI, air is now used in lower colon



CT scan without contrast showing 'apparent' density





CT scan with iodine-based contrast enhancement

Technique

- Technique refers to the factors that control image quality and patient radiation dose
- kVp (kV potential) - energy distribution of X-ray photons (recall lower energy photons are absorbed more readily
- mA - number of X-ray photons per second (controlled with tube current)
- s - gantry rotation time in seconds
- mAs - total number of photons (photons per second X seconds)
- pitch
- slice collimation
- filtration - filters placed between tube and patient to adjust energy and/or attenuation (not discussed here)

Radiation dose versus kVp

- kVp not only controls the dose but also controls other factors such as image contrast, noise and x-ray beam penetration through patient

Parameter	80 kVp	120 kVp	140 kVp
Image Contrast	***Best***	Intermediate	Poor
Noise	Most	Average	***Least***
Penetration	Least	Average	***Most***

Effective Dose Comparison with Chest PA Exam

Procedures	Eff. Dose [mSv]	Equivalent no. of chest x-rays	Approx. period of background radiation
Chest PA	0.02	1	3 days
Pelvis	0.7	35	4 months
Abdomen	1	50	6 months
CT Chest	8	400	3.6 years
CT Abdomen or Pelvis	10-20	500	4.5 years

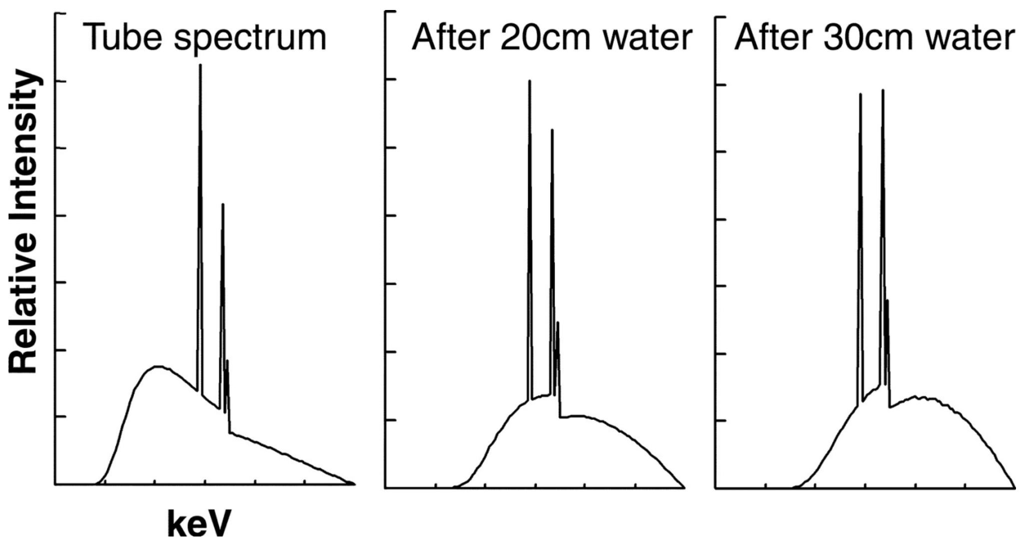
Typical Background Radiation - 3 mSv per year

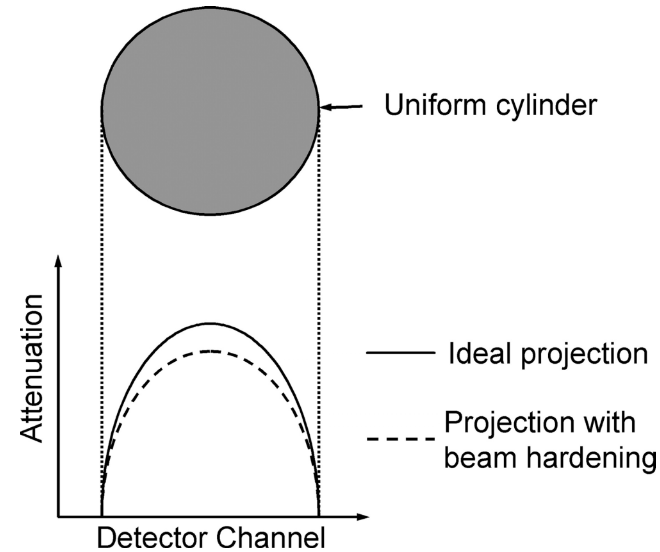
Types of CT Artifacts

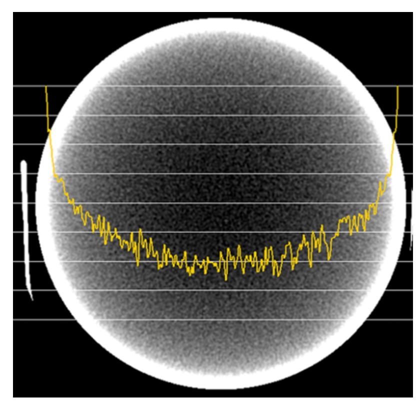
- Physics based
  - beam-hardening
  - partial volume effects
  - photon starvation
  - scatter
  - undersampling
- Scanner based
  - center-of-rotation
  - tube spitting
  - helical interpolation
  - cone-beam reconstruction
- Patient based
  - metallic or dense implants
  - motion
  - truncation

Beam Hardening

- Energy spectrum of an x-ray beam as it passes through water (rescaled)
- Mean energy increases with depth
- More photons get through, so measured attenuation is less than we would expect







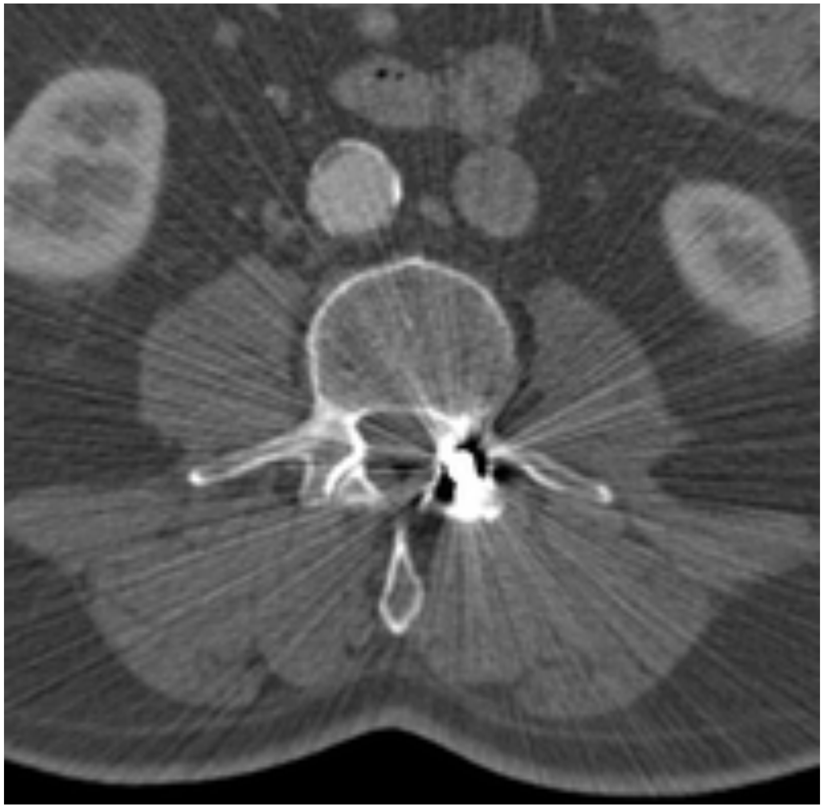
CT *image* profiles across the centre of a uniform water phantom without beam hardening correction

Beam Hardening

- If there are significant contrast changes, beam-hardening can be difficult to correct

Metallic Objects

- Occur because the density of the metal is beyond the normal range that can be handled
- Additional artifacts from beam hardening, partial volume, and aliasing are likely to compound the problem



Patient Motion

- Respiratory motion effects during helical CT scans lead to well known artifacts at the dome of the diaphragm

Truncation

- Standard CT field of view is 50 cm, but many patients exceed this
- Not often a problem for CT, but can be a problem when a truncated CT is used for PET attenuation correction