Some reference material

Physics reference book on medical imaging:

- A good one is “The Essential Physics of Medical Imaging”, 3rd Ed. by Bushberg et al. ($170! new). However, there are several similar books by Hendee, Huda, etc.. e.g. Search "Physics Medical Imaging” online.

Articles on risks for X-ray and CT imaging:

- Cancer risks from CT scans: now we have data, what next? Brenner and Hall. Radiology (2012) vol. 265 (2) pp. 330-1

- Radiation risks of medical imaging: separating fact from fantasy. Hendee and O'Connor. Radiology (2012) vol. 264 (2) pp. 312-21

Lecture-3 Homework Questions Feedback

- **Are there 2D bow-tie filters?**

→Yes, for radiography systems. Sometimes they are even customized.

- **What is the difference between contrast and signal?**

→Contrast is a signal ratio

- **What is the purpose of a phosphor screen in front of X-ray film?**

→X-ray film is maybe a misnomer. Film has very low X-ray stopping power and is more sensitive to visible light. Phosphor or scintillators are used to convert x-rays to optical energies so we can use photosensors to convert to an electronic signal. Semiconductor detectors that detect x-rays are also used, and typically have better energy resolution, but are less attenuating and often cost more.

Lecture-3 Homework Questions Feedback

- **Does inherent quantum noise mean that we can never truly get a perfect signal to noise ratio?**

→Counting events that occur at random intervals during a fixed integration time results in a random variable. If events are uncorrelated, the number we count is a Poisson random variable.

Who here is familiar with the Poisson distribution?

So, our signal is limited by our integration time and the mean number of events we get in each image voxel.

Lecture-3 Homework Questions Feedback

- **What value of SNR is good enough for the different types of scans and their respective applications?**

→SNR requirements depend on how distinct two classes we are trying to distinguish are. A good separation would be 3-sigma st.dev. between "signal present" and "signal absent" cases. However, you must truly answer how far apart is enough by doing a risk assessment for the type of task you are performing. So, you first need to specify what your imaging task is. Then you can set requirements on every part of our imaging chain (e.g., radio-tracer specificity, biology, administration of tracer, hardware, acquisition time, reconstruction options, etc). For example, if we want to detect a lesions that are at least 1-cm in size, then we need enough counts in the region of the lesion to distinguish it from background. This will depend on how specific the tracer is taken up in the lesion vs. background. And it will depend on how well we resolve the lesion (partial volume effects due to resolution will bleed counts into the background). This topic of determining requirements for the entire imaging decision chain is called "Objective Assessment of image Quality".

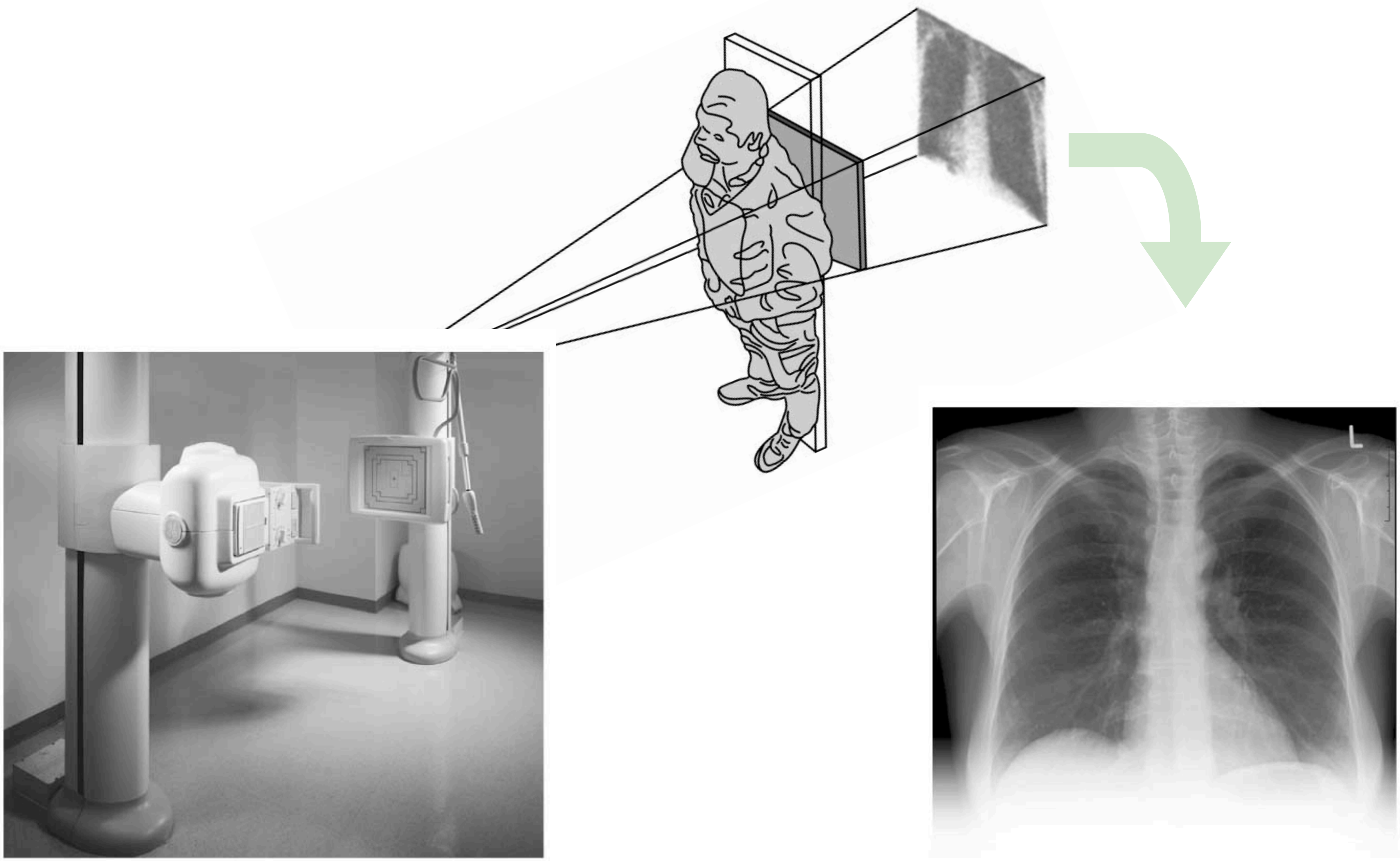
Lecture-3 Homework Questions Feedback

- **How is the imaging equation used?**

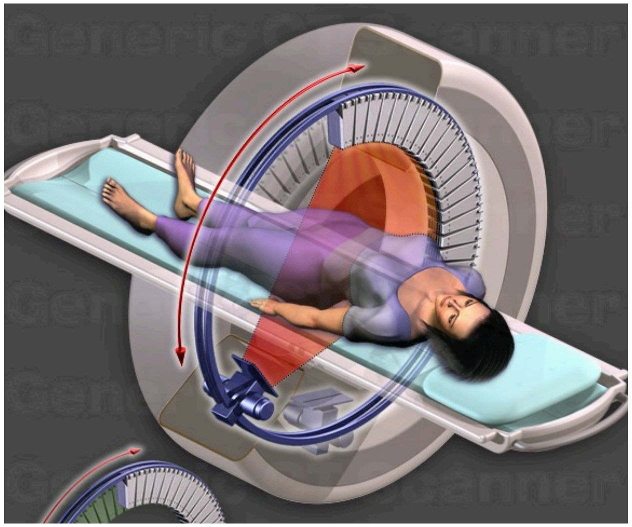
→The imaging equations represents the forward problem in how we obtain signals with our imaging system. We can then use the imaging equation to correct for imaging system effects so we can measure a quantity related to the object we are imaging. For example in 2D radiography, inversion of our image equation allows us to correct for geometry, sensitivity, and other system specific parameters that affect our estimate of the integrated attenuation through the subject. For 3D image reconstruction, the problem of inverting the imaging equation to do a tomographic reconstruction is a bit more complicated, but is still tractable using either analytic or iterative numerical methods.

X-ray Computed Tomography

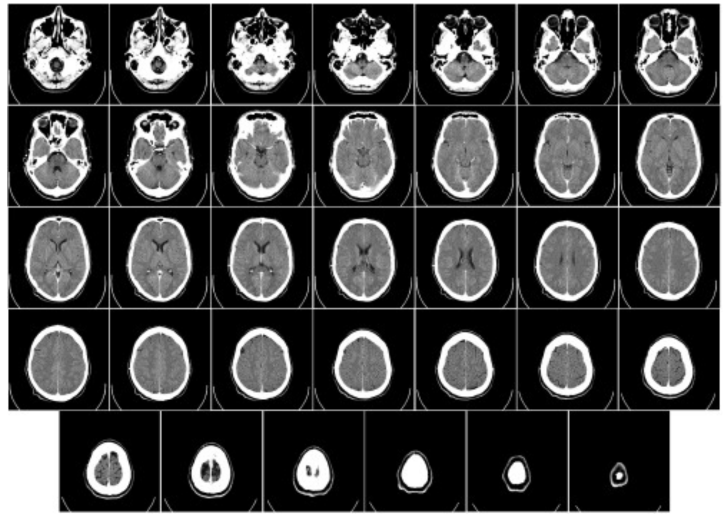
Types of Images: Projection Imaging



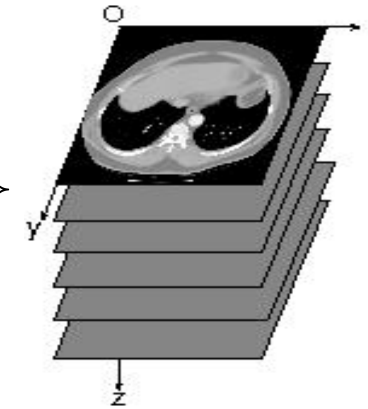
Types of Images: Tomography Imaging



tomographic acquisition

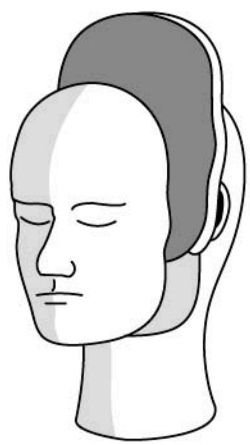


reconstruction of multiple images

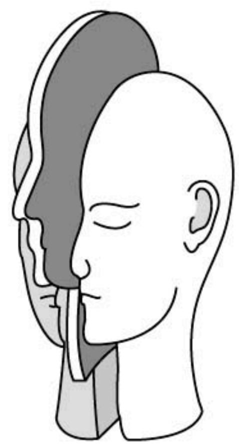


form image volume

transaxial or axial view



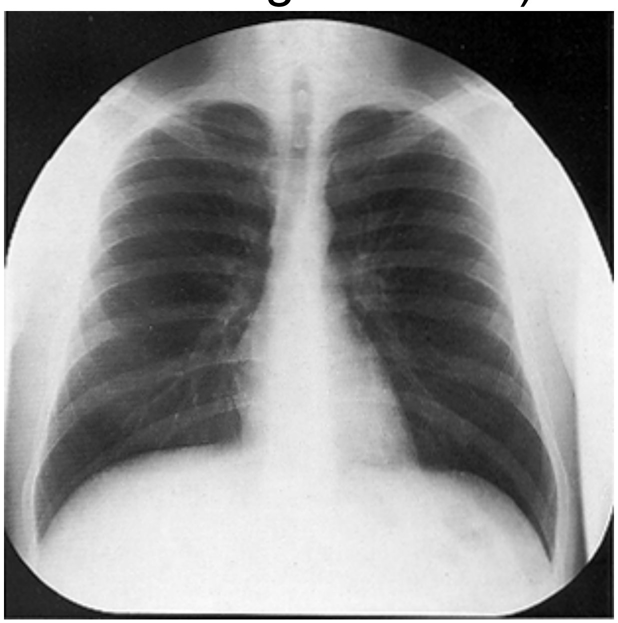
coronal view



sagittal view

Comparing Projection and Tomographic Images

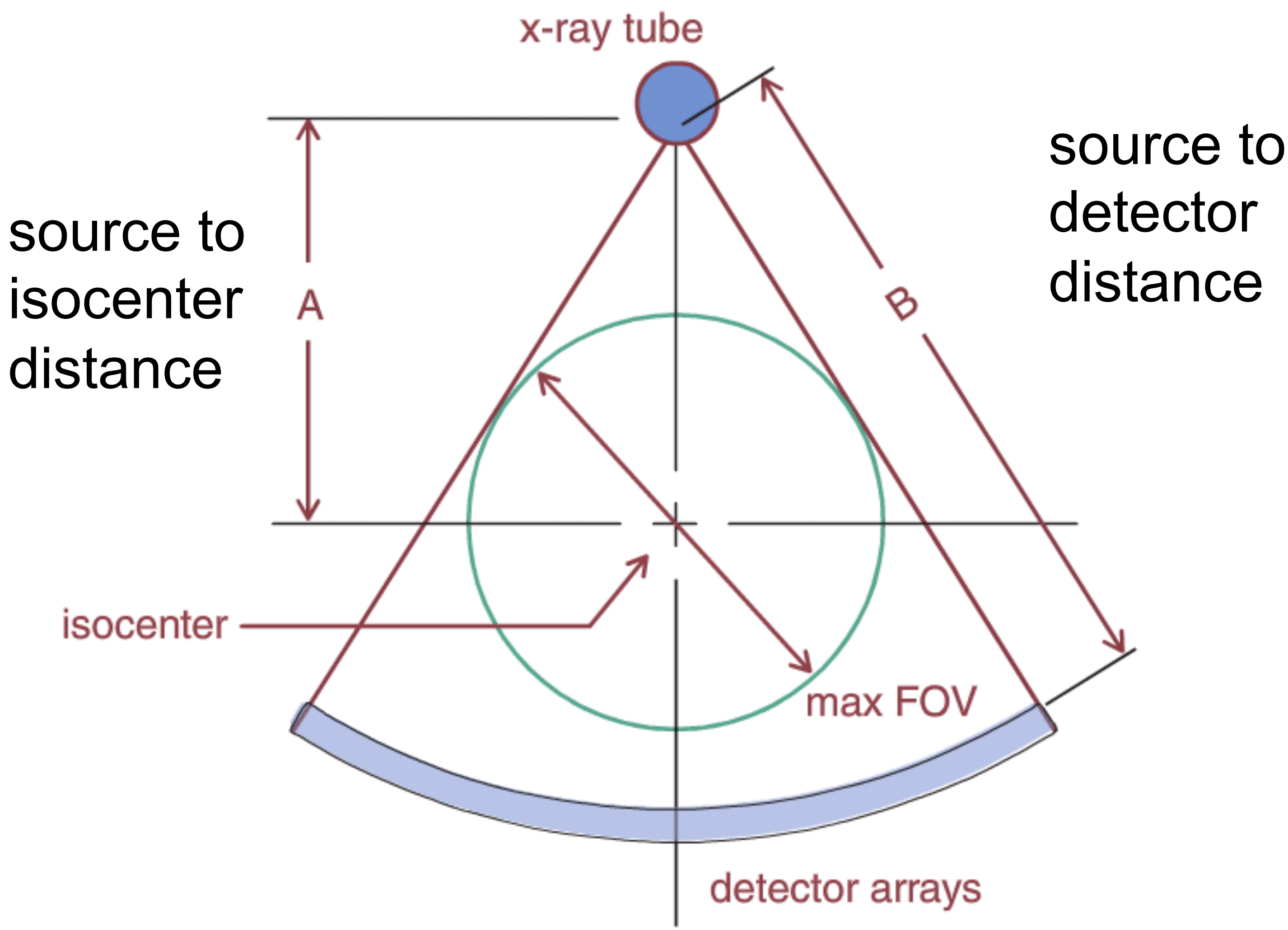
- Hounsfield's insight was that by imaging all the way around a patient we should have enough information to form a cross-sectional image
- *Sir* Godfrey Hounsfield shared the 1979 Nobel Prize with Allan Cormack (of FBP fame), funded by the EMI and the Beatles
- Radiographs typically have higher resolution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



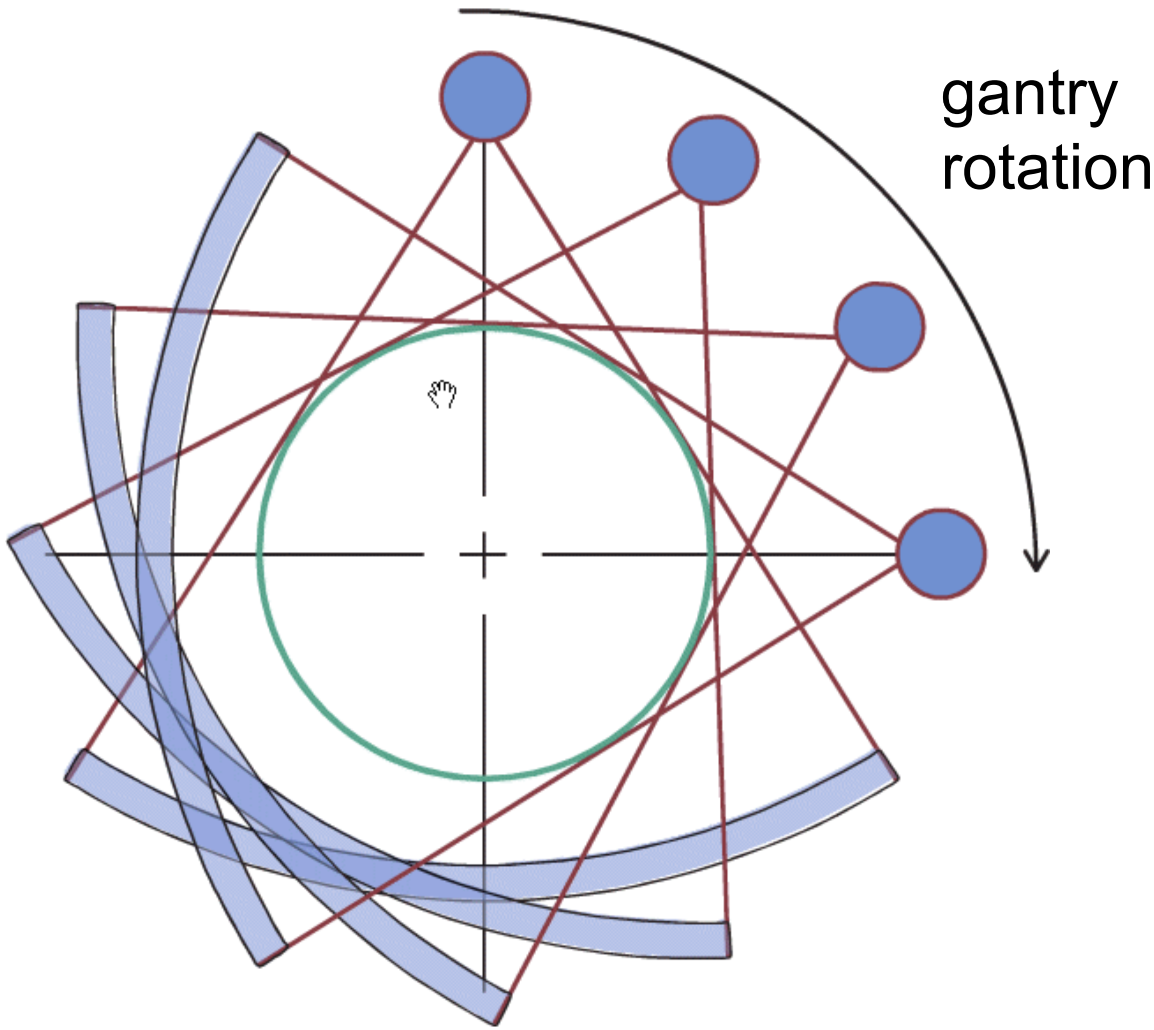
Chest radiograph

Coronal *section* of a 3D CT image volume

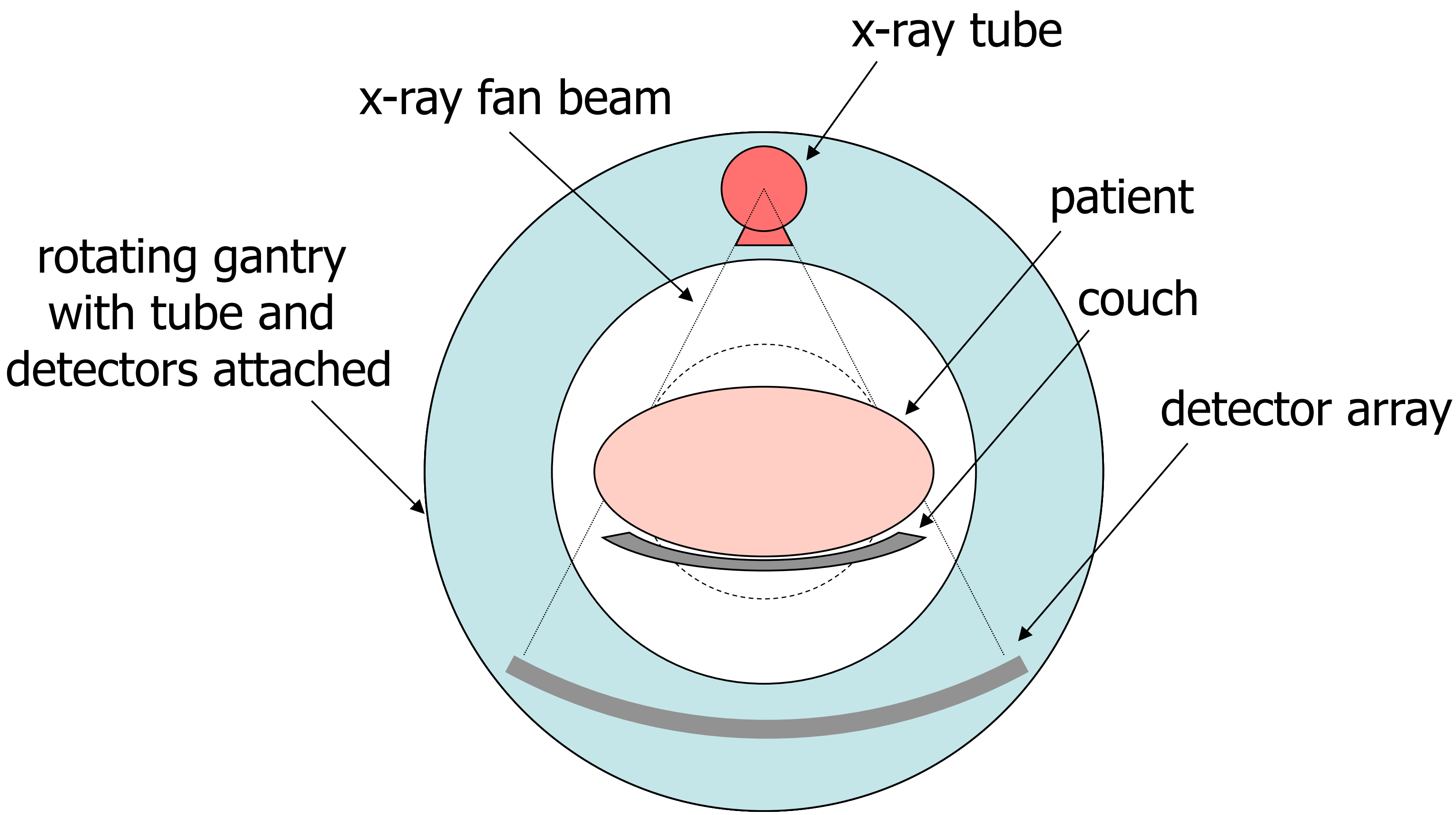
CT Scanner Geometry



CT Scanner Geometry

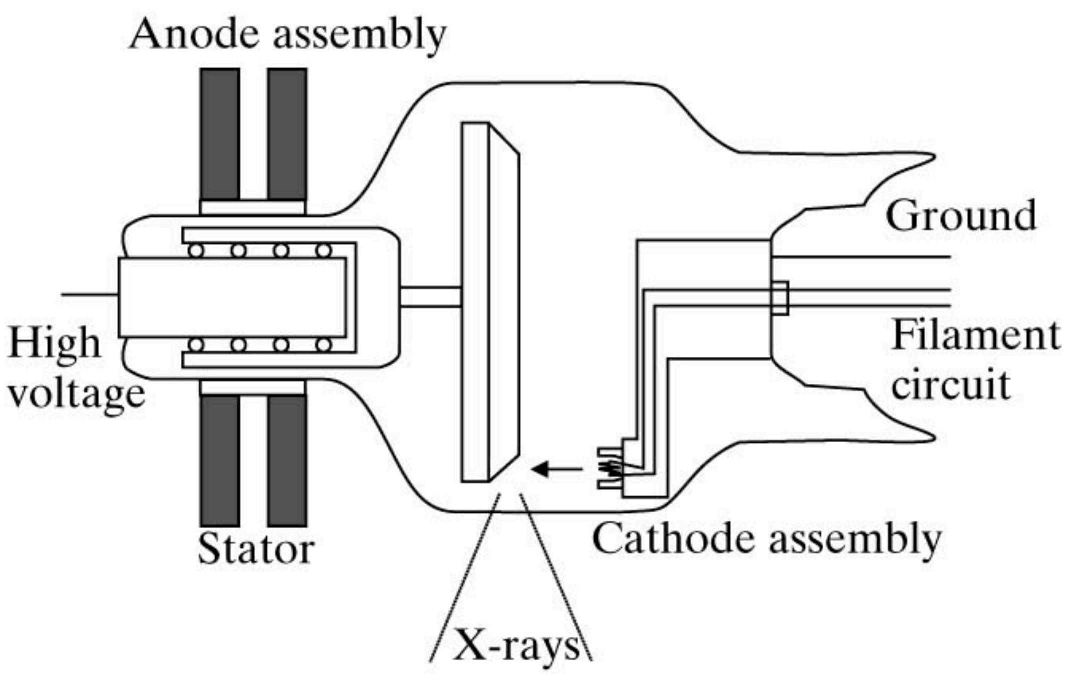


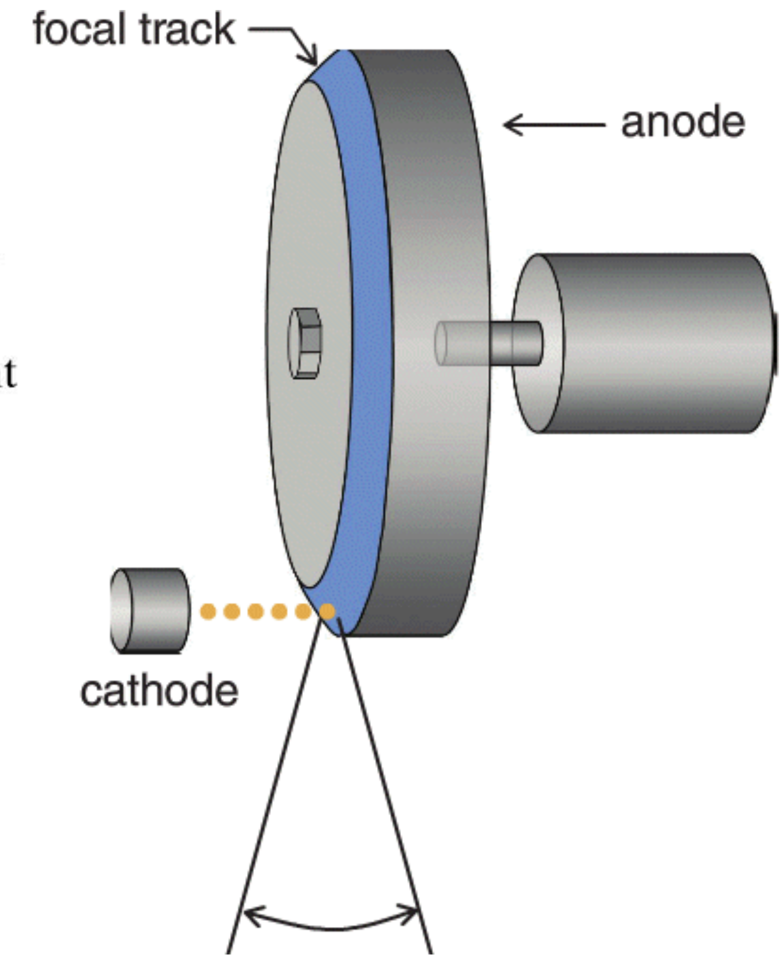
CT Scanner Components



- 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

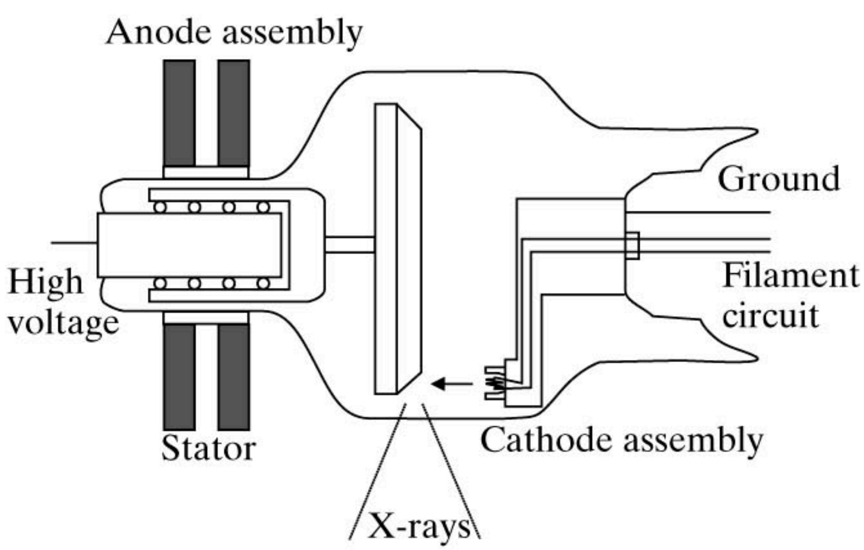


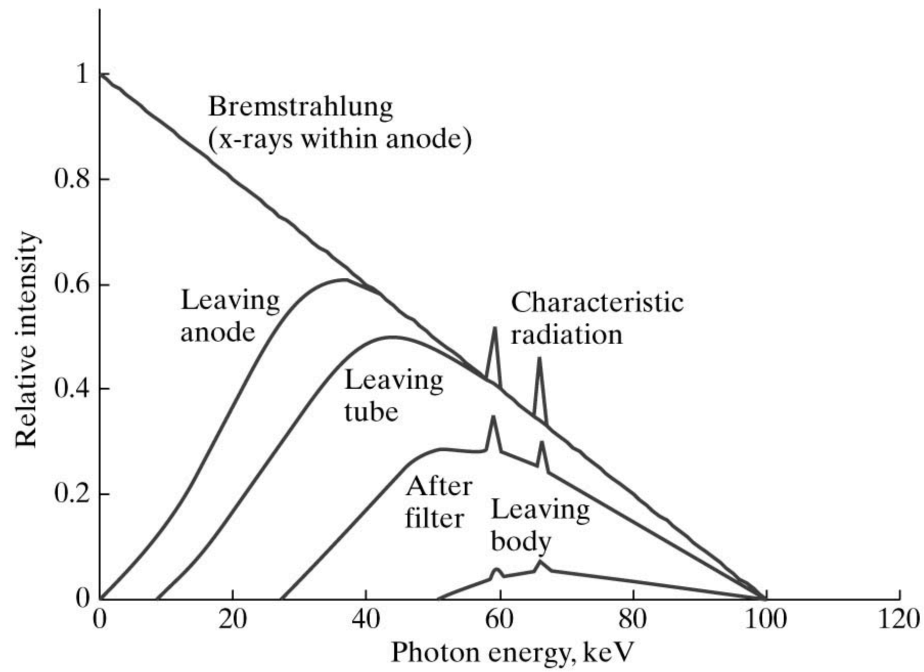


- In a vacuum assembly
- A resistive filament is used to 'boil off' electrons in the cathode with a carefully controlled current (10 to 500 mA)
- Free electrons are accelerated by the high voltage towards the anode

X-ray tubes

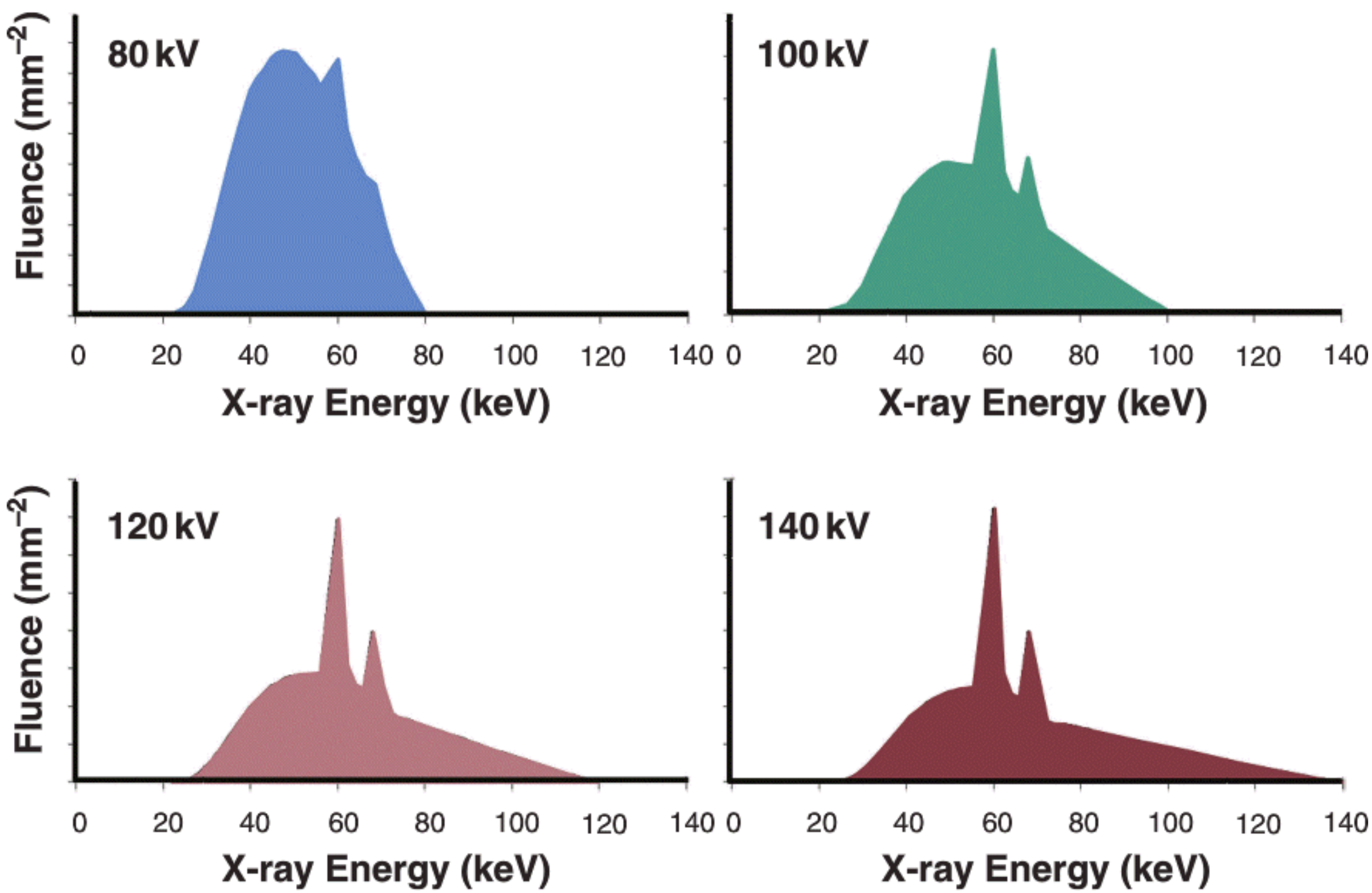
- Voltage determines maximum and x-ray energy, so is called the kVp (i.e. kilo-voltage potential), typically 90 kVp to 140 kVp for CT
- High-energy electrons smash into the anode
  - More than 99% energy goes into heat, so anode is rotated for cooling (3000+ RPM)
  - Bremmstrahlung then produces polyenergetic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



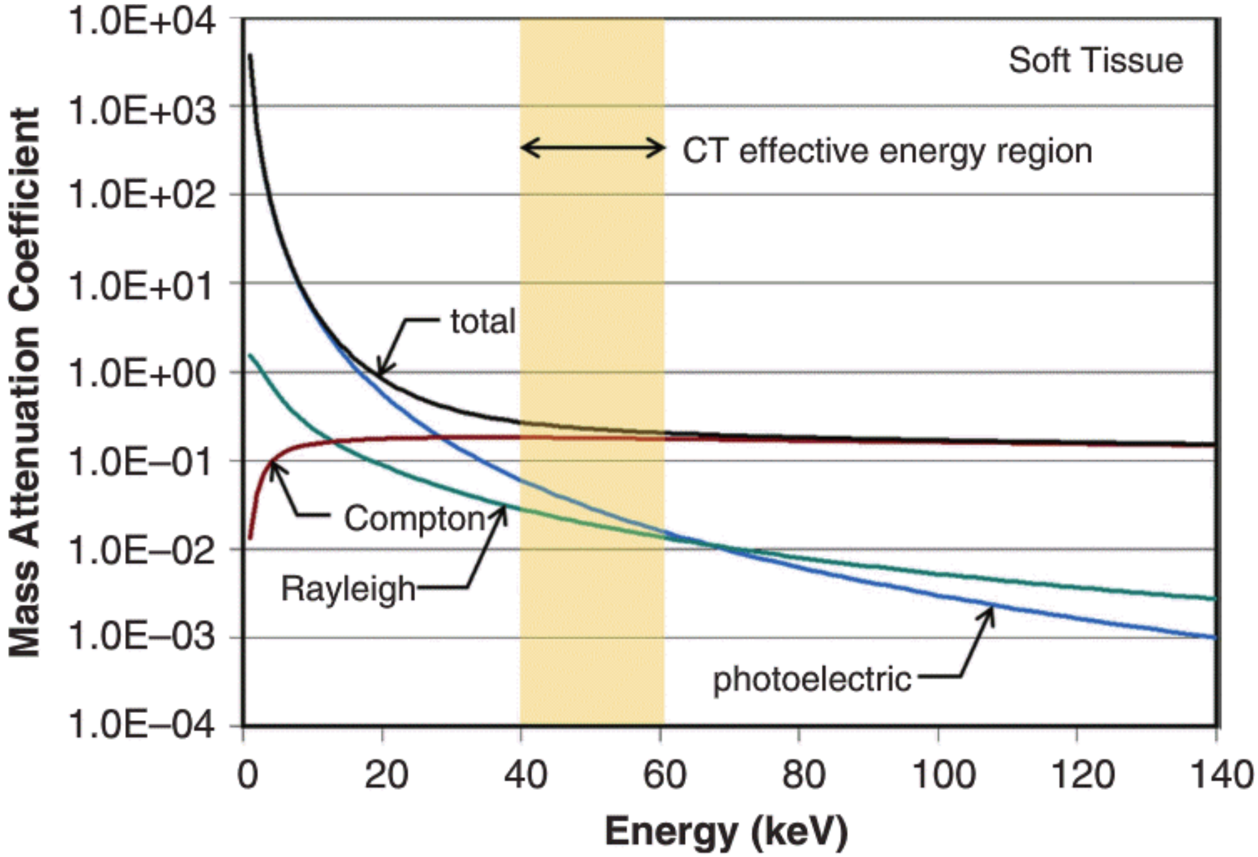


Typical X-ray spectra in CT

scaled to peak fluence



Mass attenuation coefficient versus energy



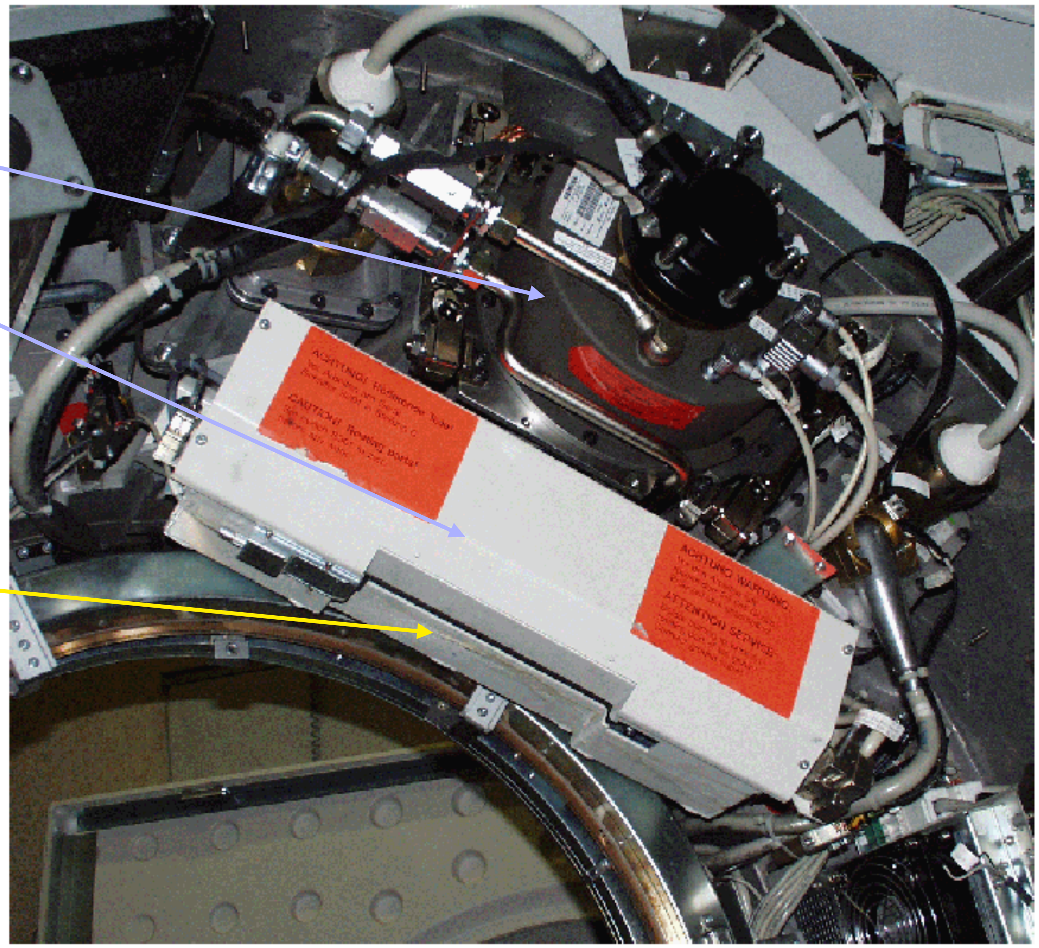
Pre-Patient Collimation

- Controls patient radiation exposure

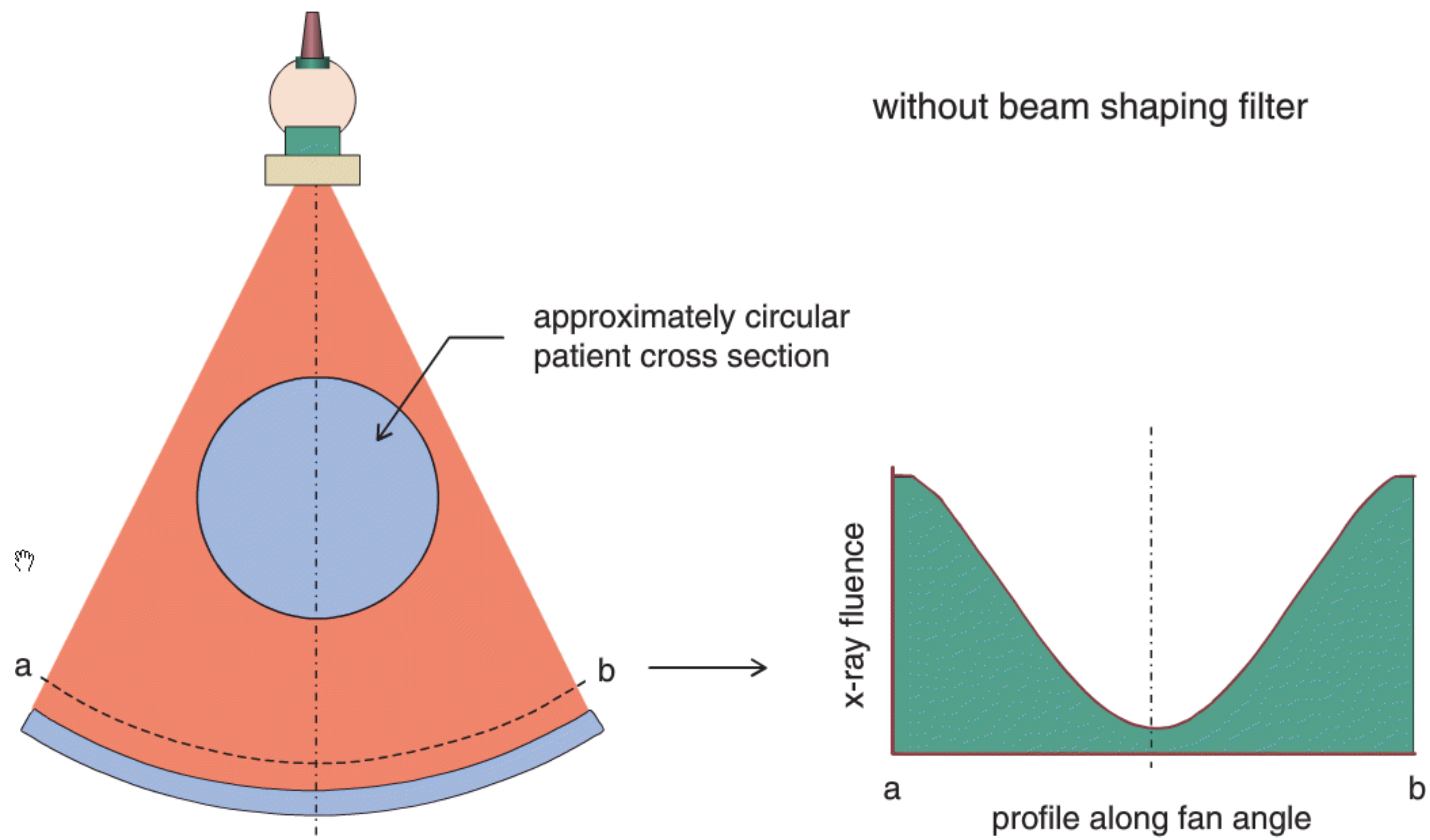
X-ray tube

collimator and filtration assembly

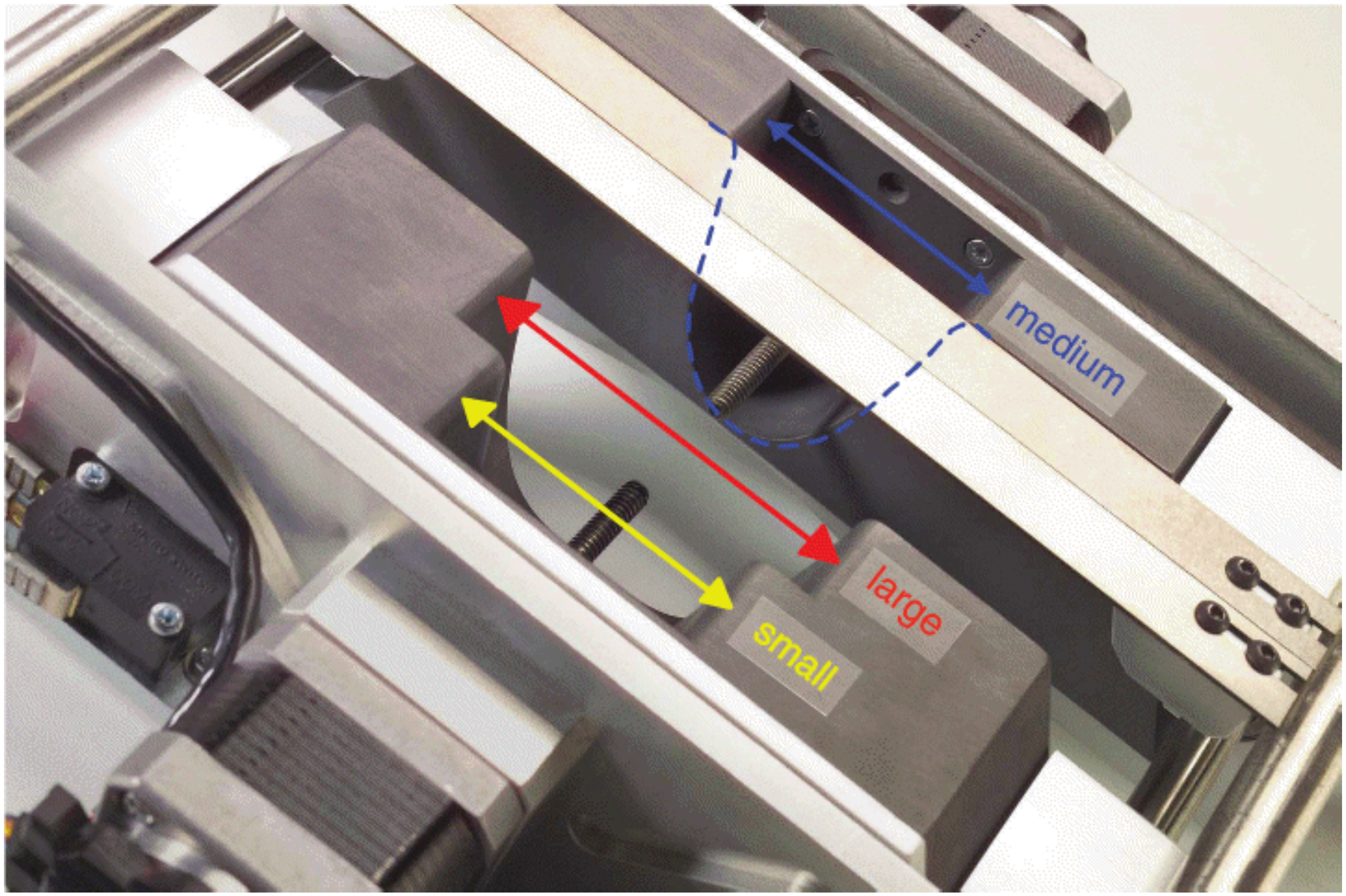
X-ray slit



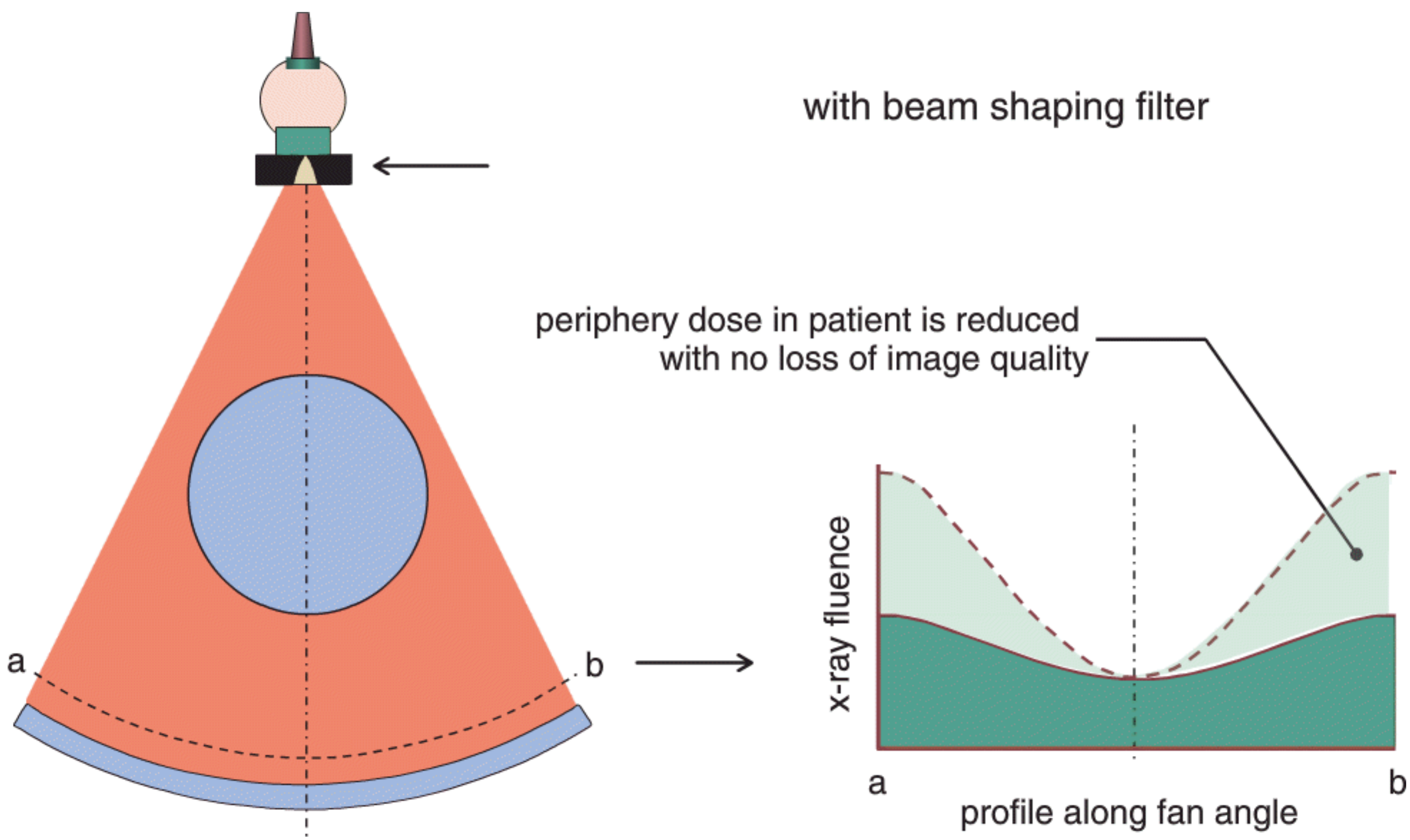
Need for x-ray beam shaping



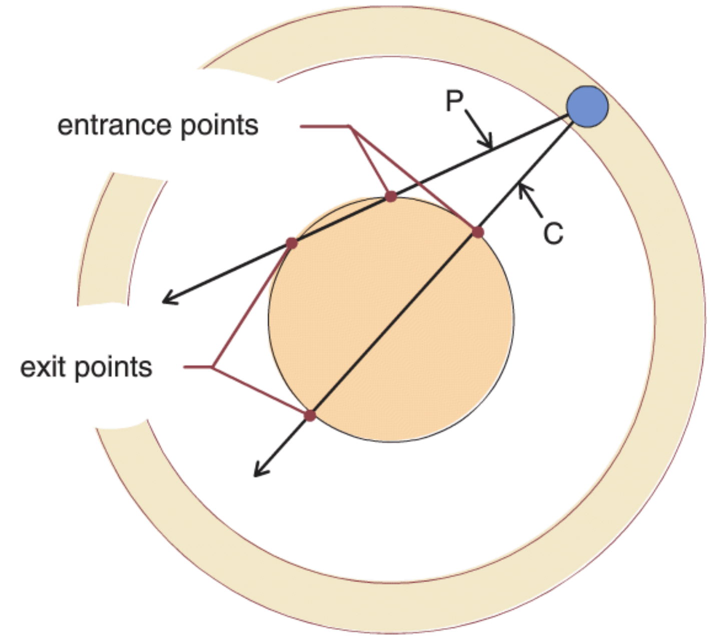
Addition of 'bow-tie' filters for beam shaping

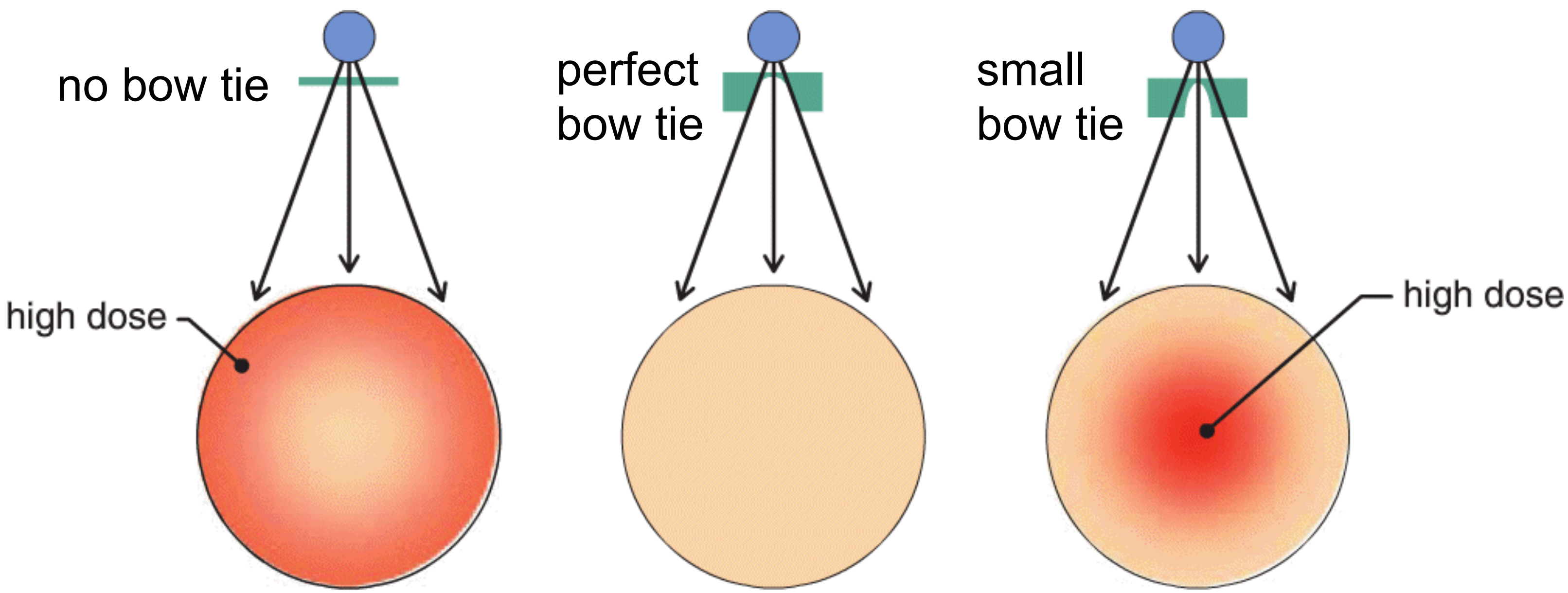


Use of 'Bow-tie' beam shaping



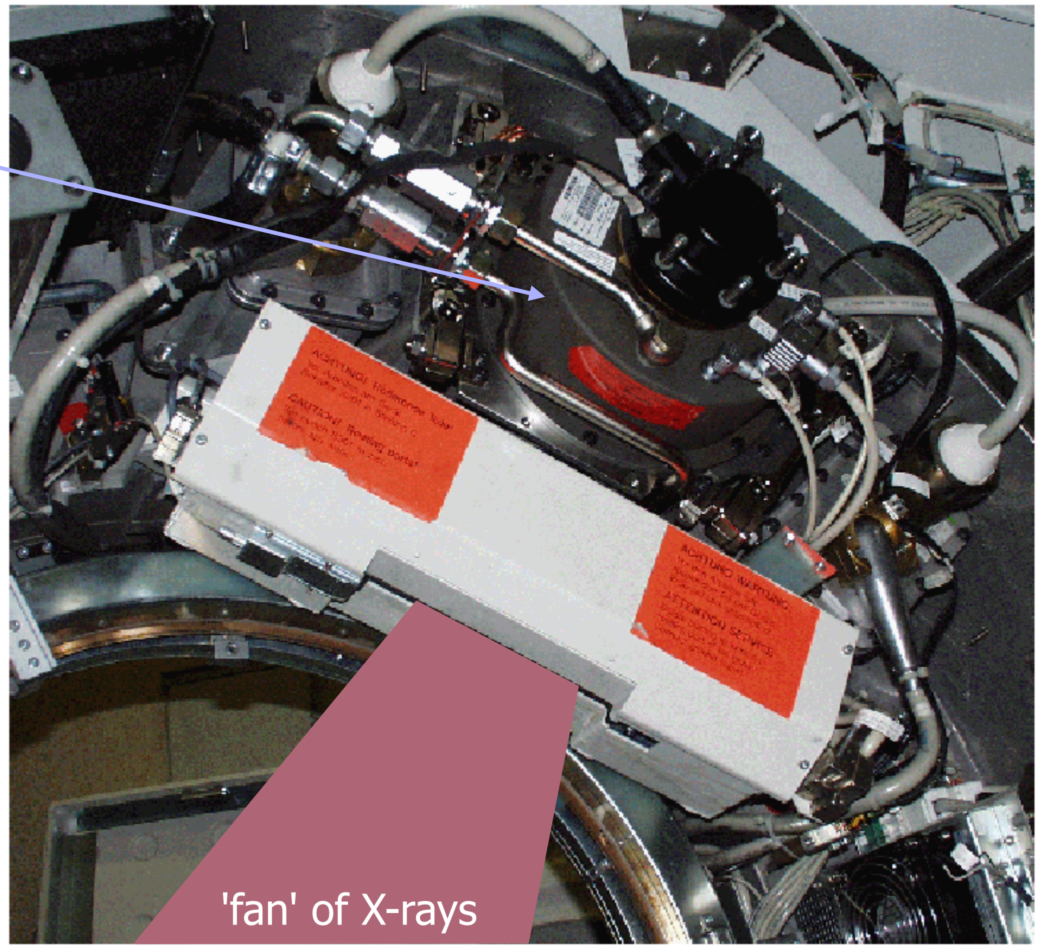
Radiation dose considerations



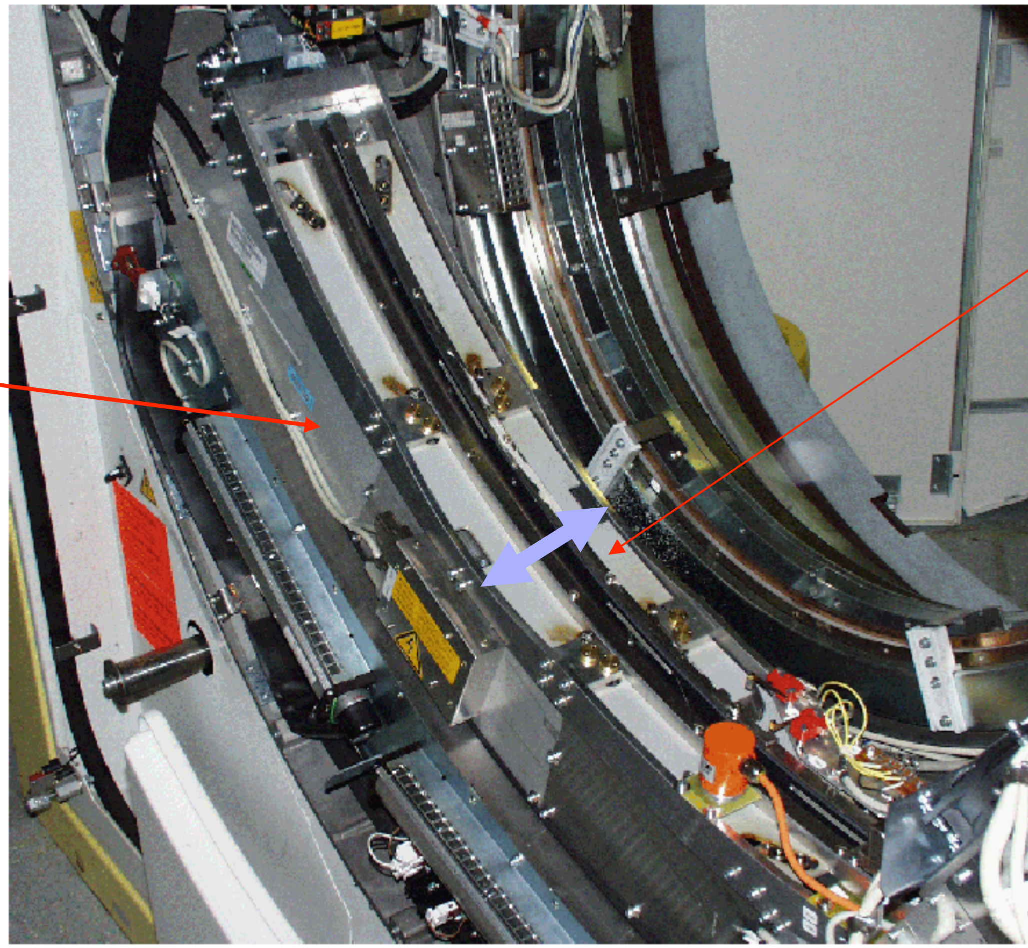


Pre-Patient Collimation

- Controls patient radiation exposure

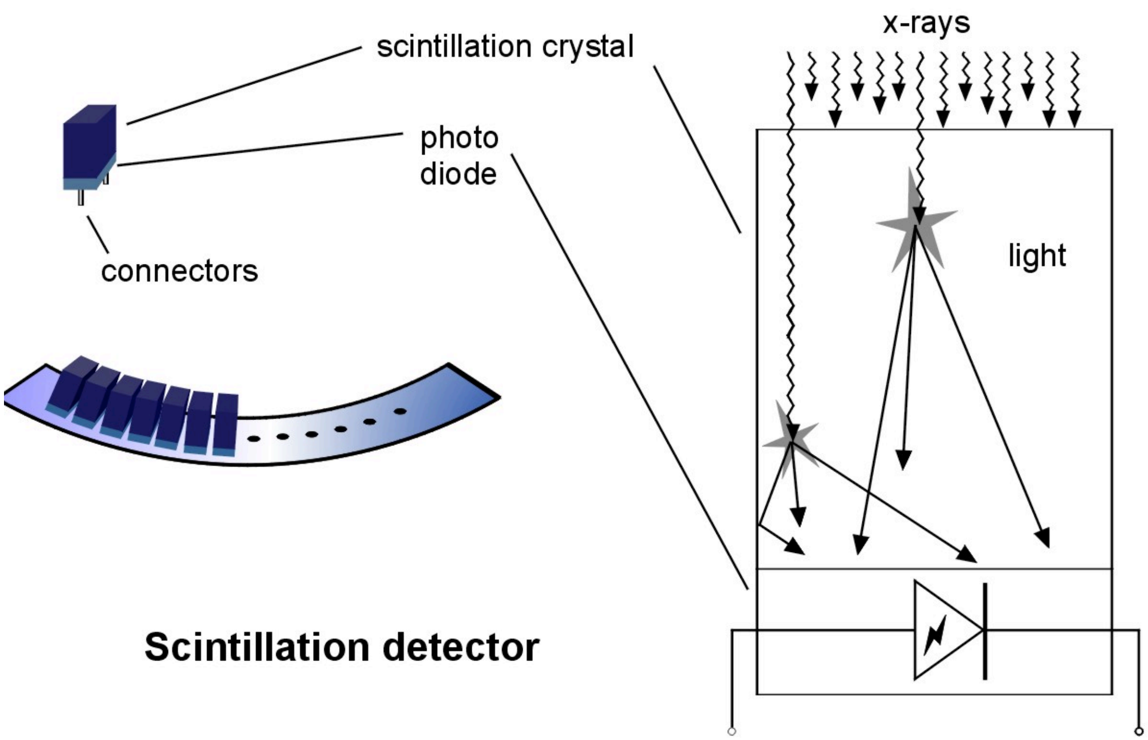


X-ray Detector Assembly



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

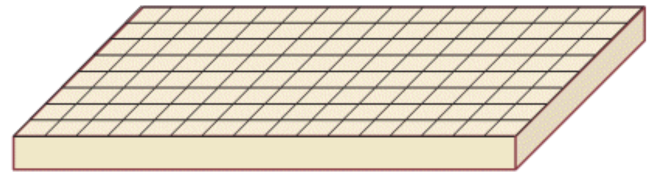


Scintillation detector

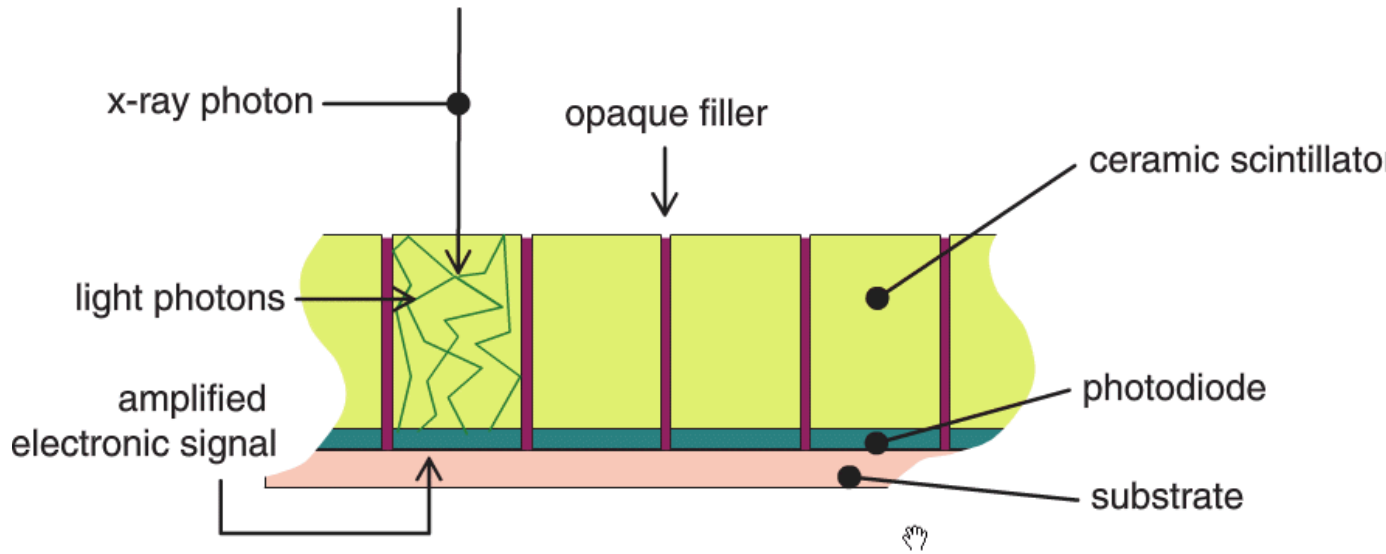
X-ray CT Detectors

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

Sintered to increase density



perspective view

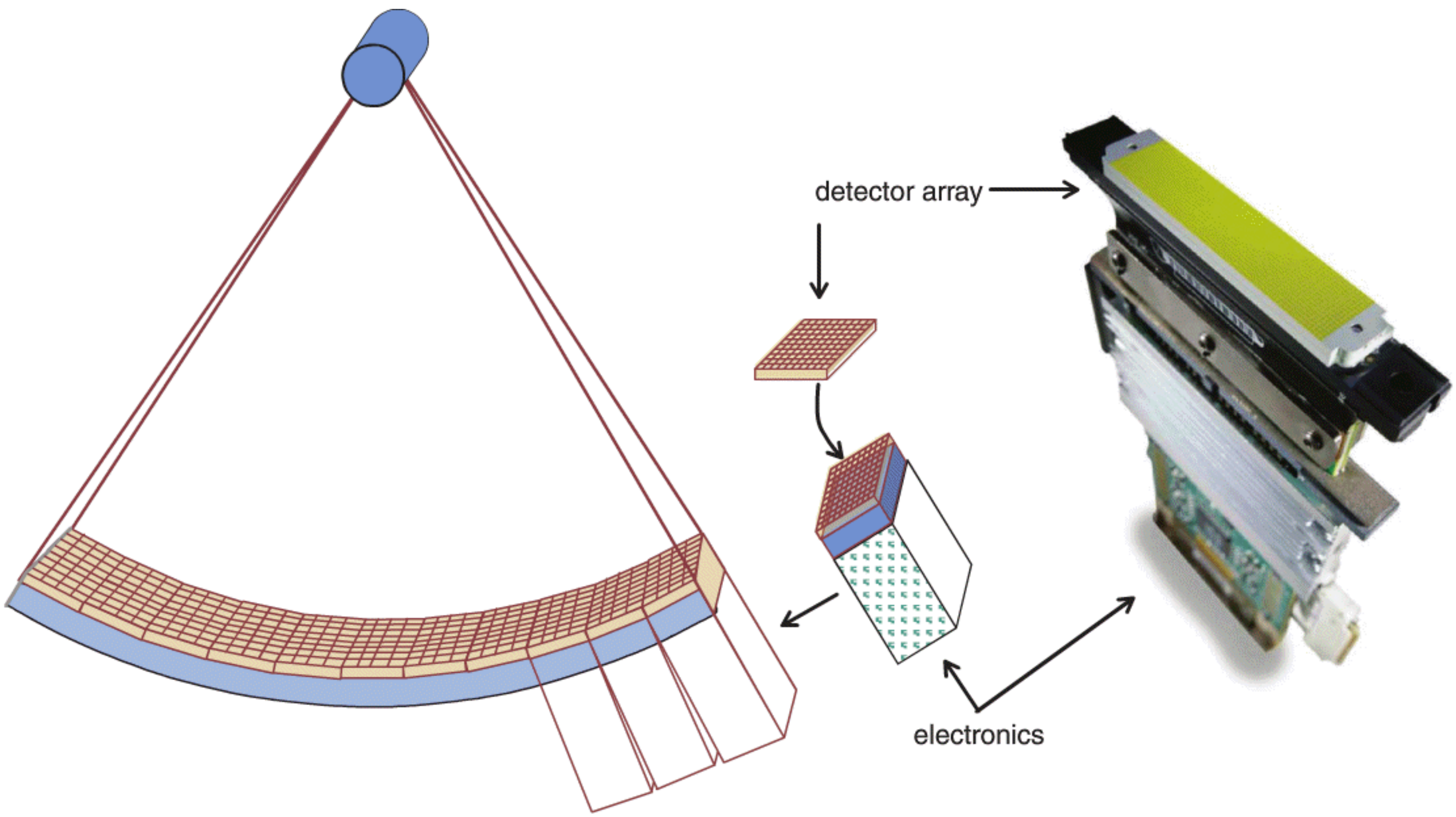


side view

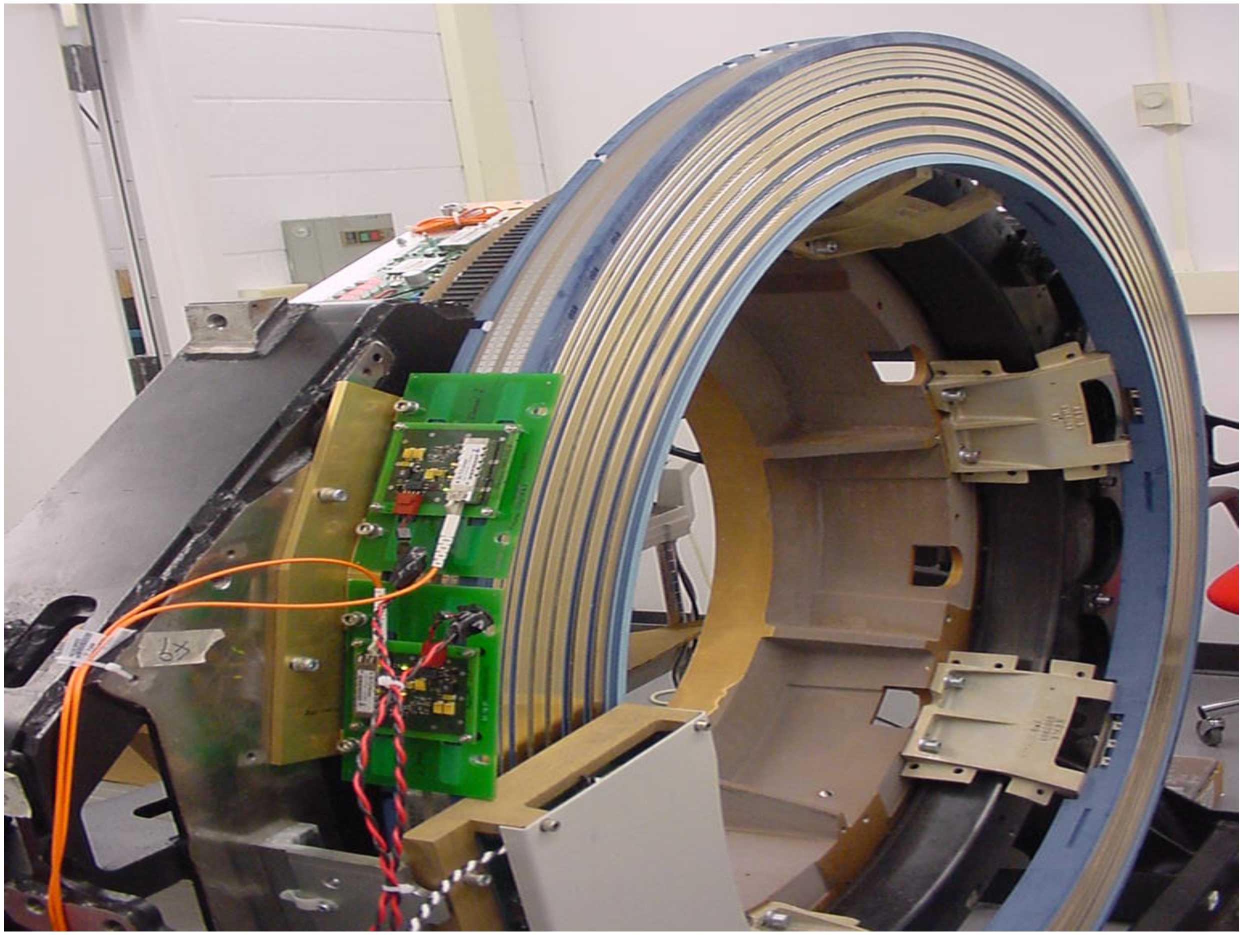
X-ray CT Detectors

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

e.g.,

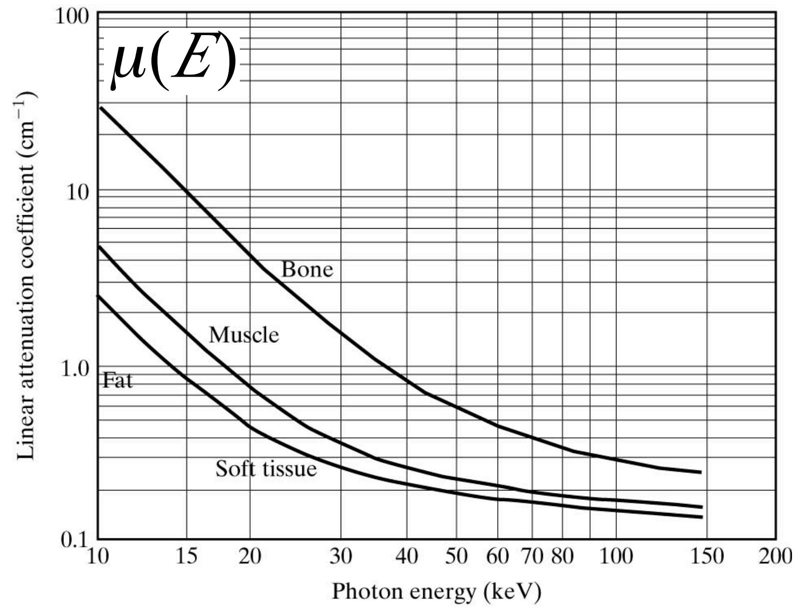
https://www.youtube.com/watch?v=2CWpZKuy-NE

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

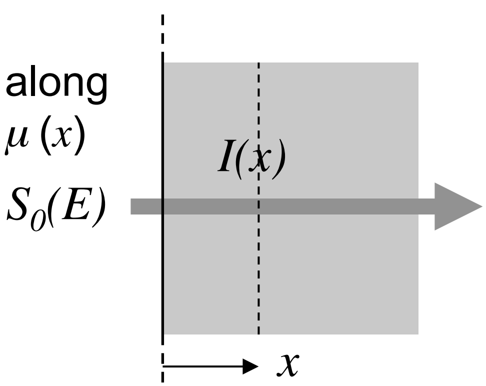
Narrow-beam Polyenergetic Attenuation

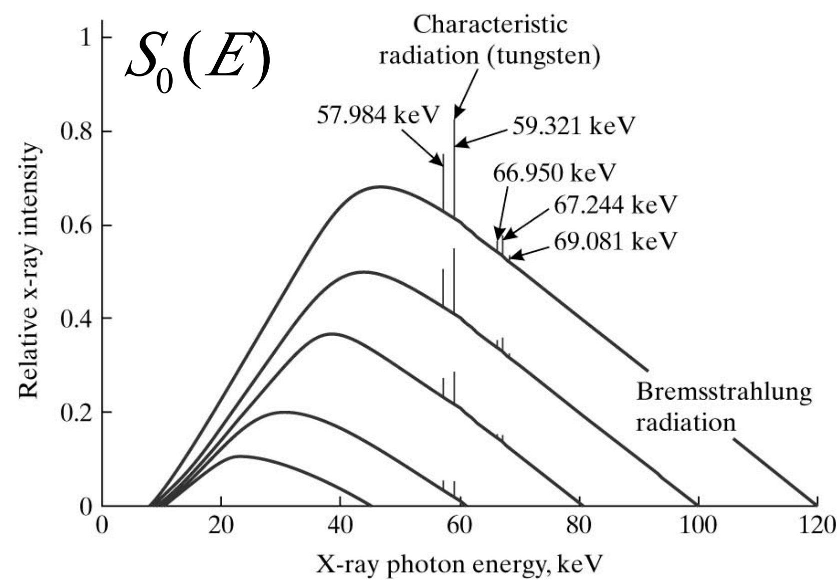
- The attenuation depends on material (thus position of material) and energy
- With bremsstrahlung radiation, there is a weighted distribution of energies
- We combine previous results to get the imaging equation

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



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*

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

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 $g_d \triangleq -\ln\left(\frac{I_d}{I_0}\right)$

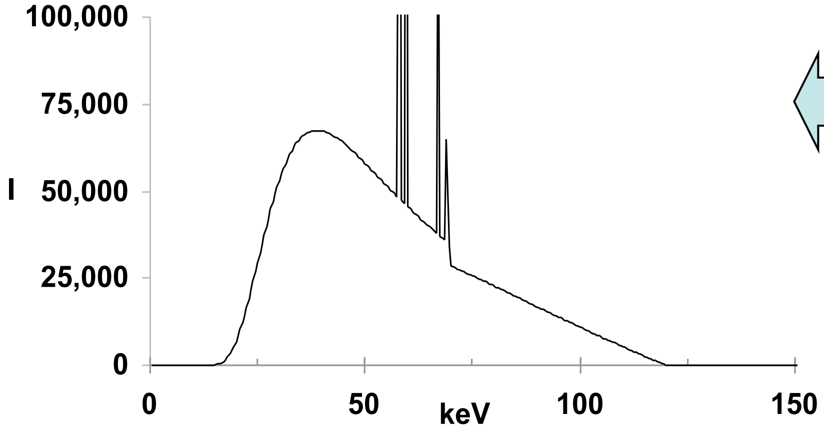
- Giving an x-ray transform $g_d = -\int_0^d \mu(s,\bar{E})\,ds$

  (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

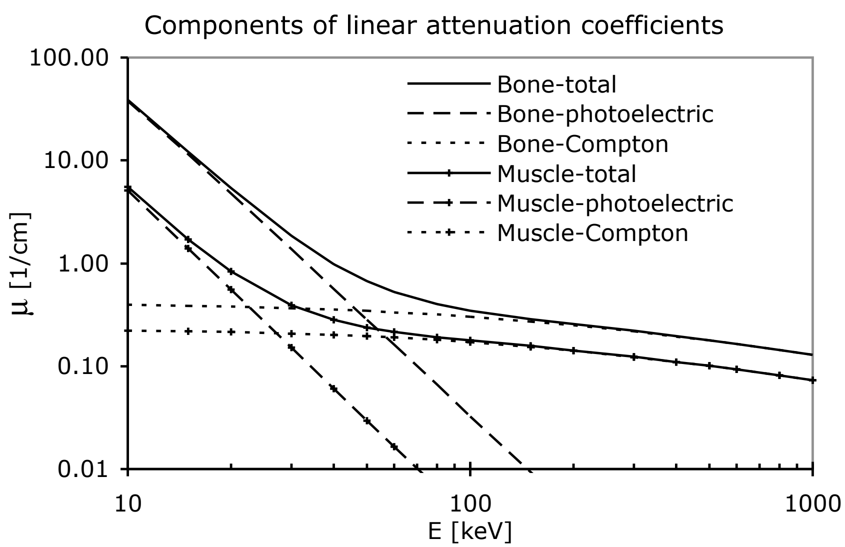
X-ray CT Image Values

- 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

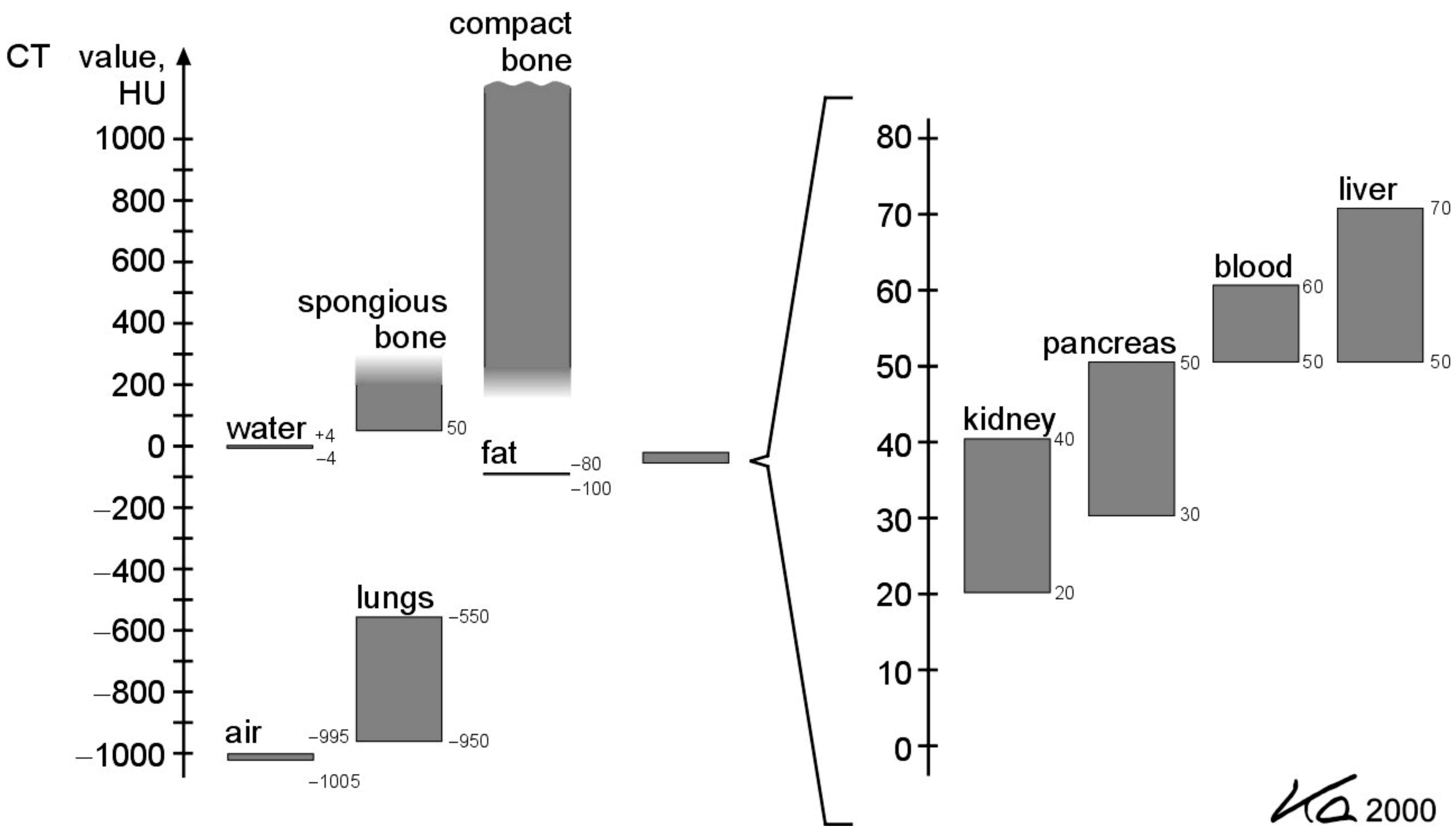
- We can't solve the real inverse problem since we have a mix of densities of materials, each with different Compton and photoelectric attenuation factors at different energies, and a weighted energy spectrum
- The best we can do is to use an *ad hoc* image scaling
- The CT number for each pixel, (x,y)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

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

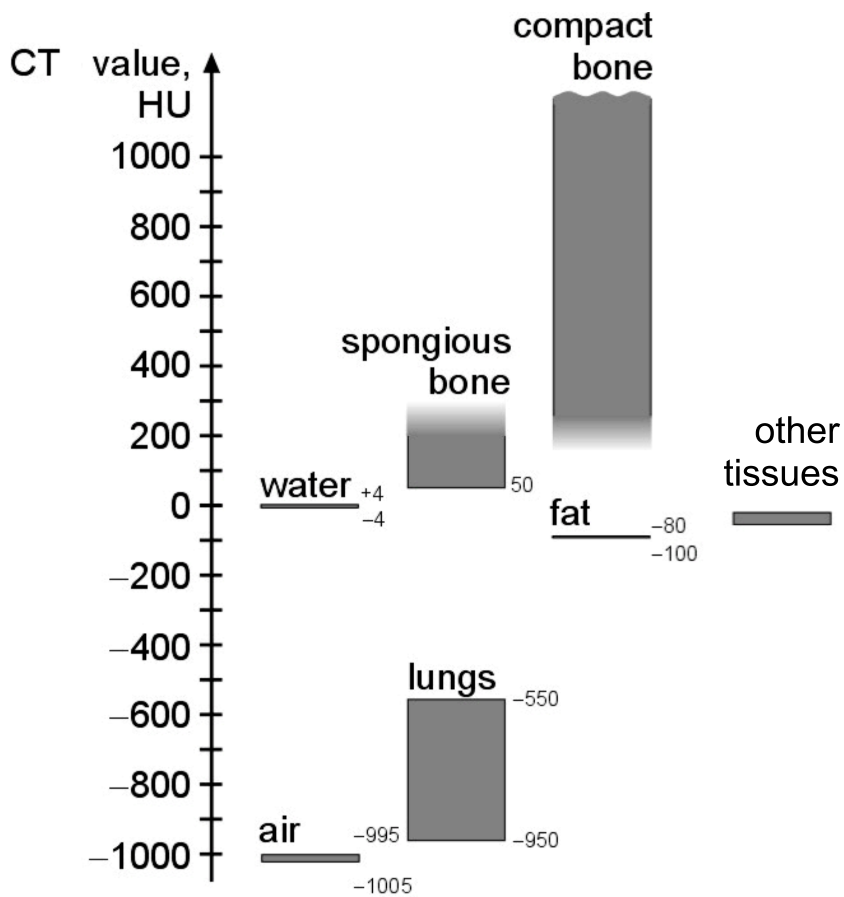
- $\mu(x, y)$ is the reconstructed attenuation coefficient for the voxel, μ_{water} is the attenuation coefficient of water and CT(x,y) is the CT number (using *Hounsfield units*) of the voxel values in the CT image

CT Numbers

- Typical values in Hounsfield Units

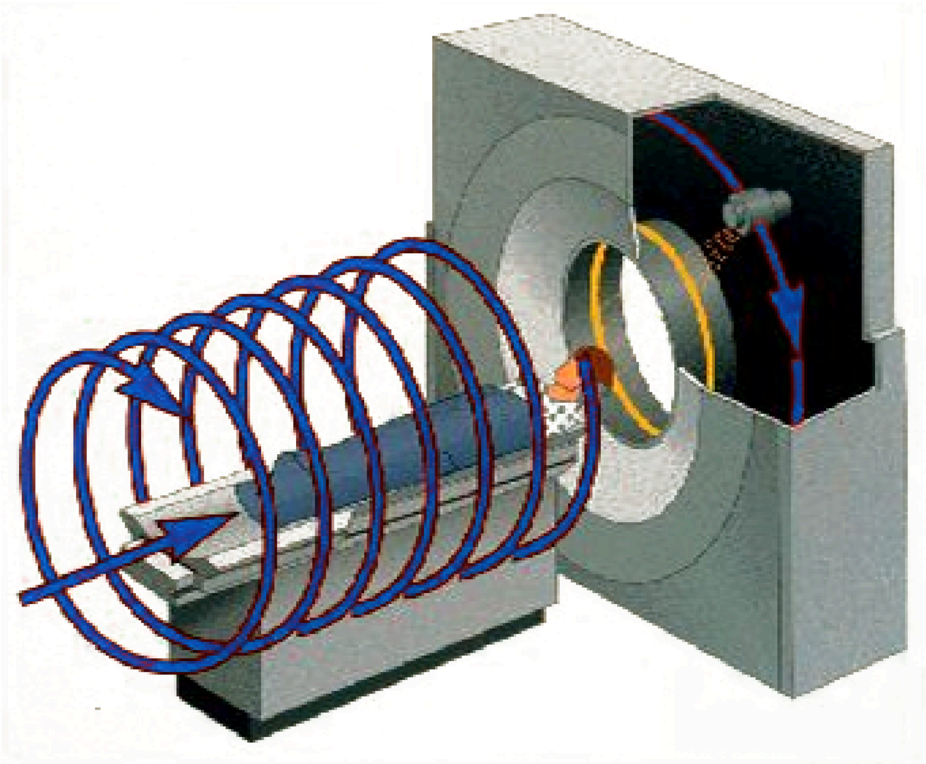


CT scan showing 'apparent' density



Helical CT Scanning

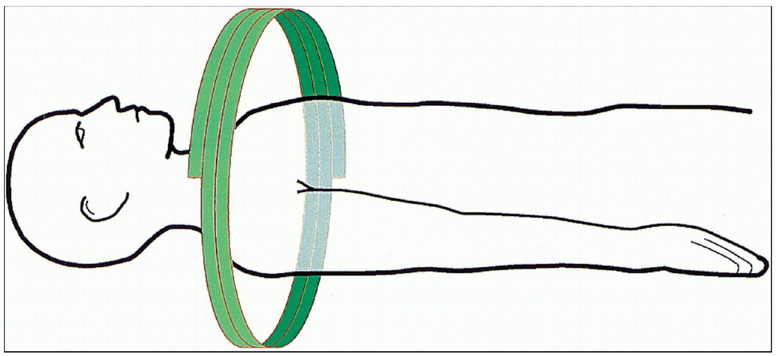
- The patient is transported continuously through gantry while data are acquired continuously during several 360-deg rotations
- The ability to rapidly cover a large volume in a single-breath hold eliminates respiratory misregistration and reduces the volume of intravenous contrast required



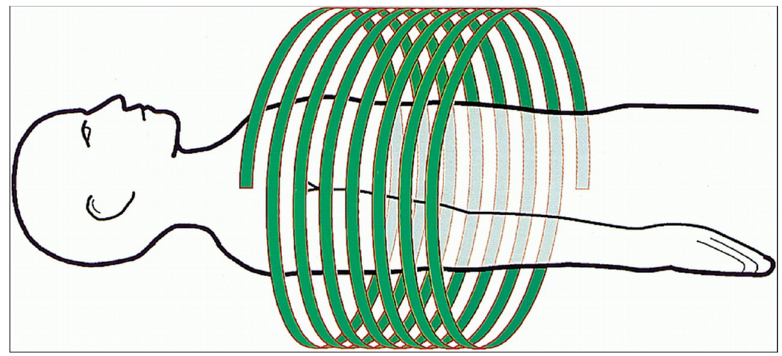
Pitch

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

slingle slice example



Pitch = 1

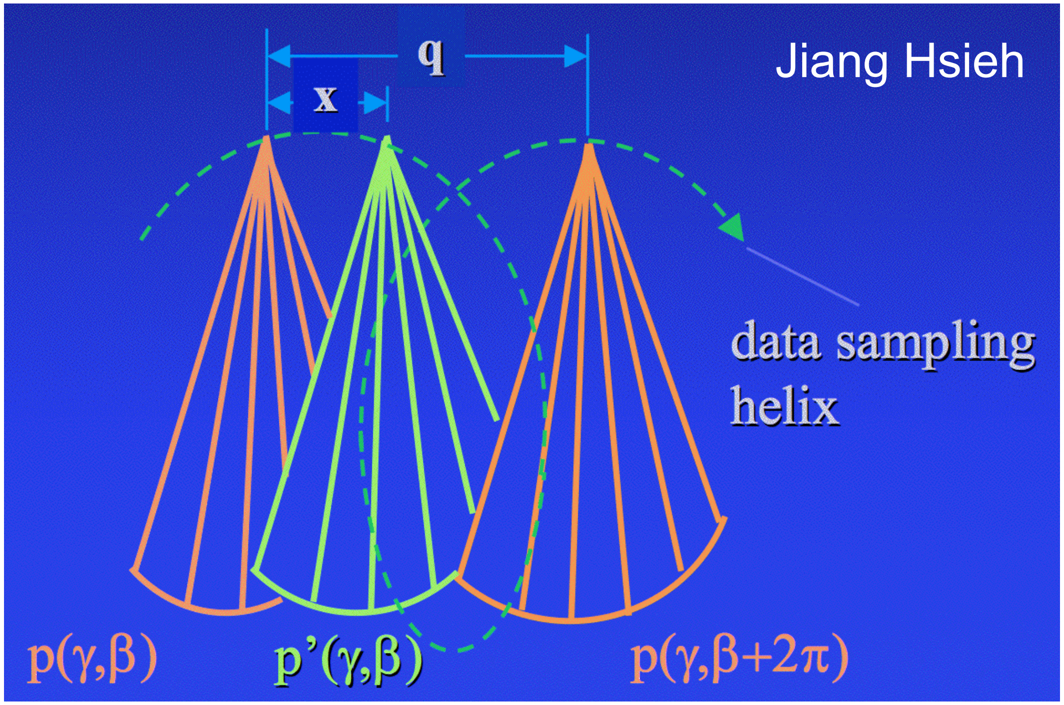


Pitch = 2

- 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

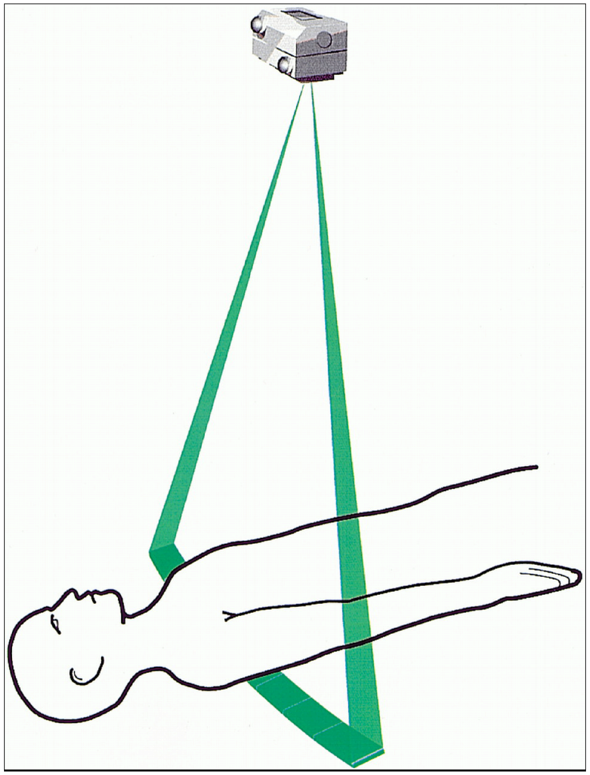


$$p'(\gamma,\beta) = wp(\gamma,\beta) + (1-w)p(\gamma,\beta+2\pi)$$

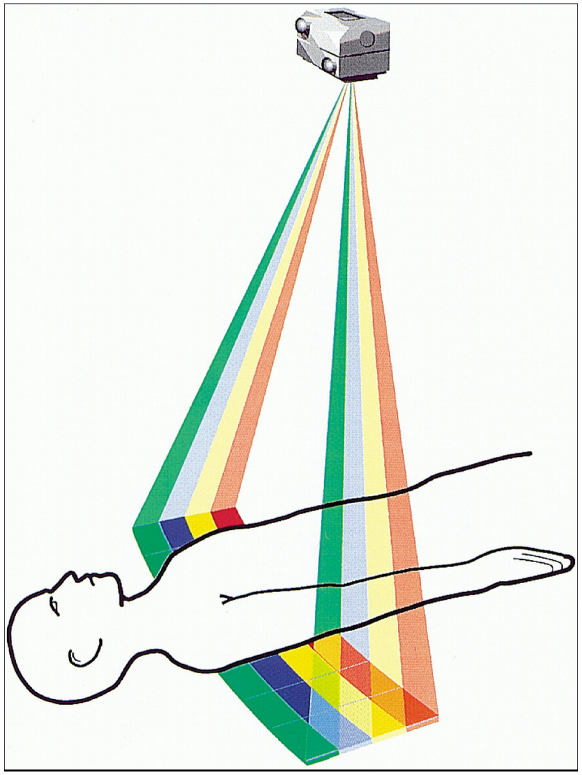
where $w = (q-x)/q$

Single versus Multi-row Detectors

- Can image multiple planes at once



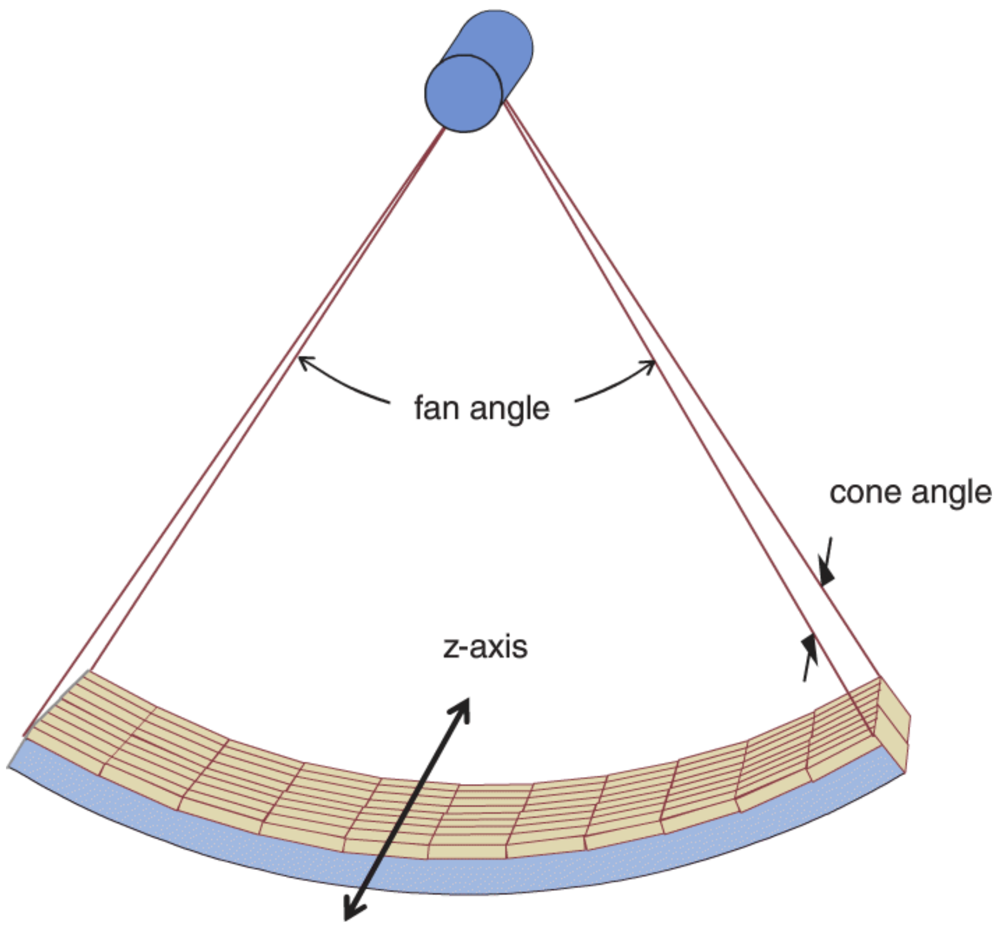
1 detector row



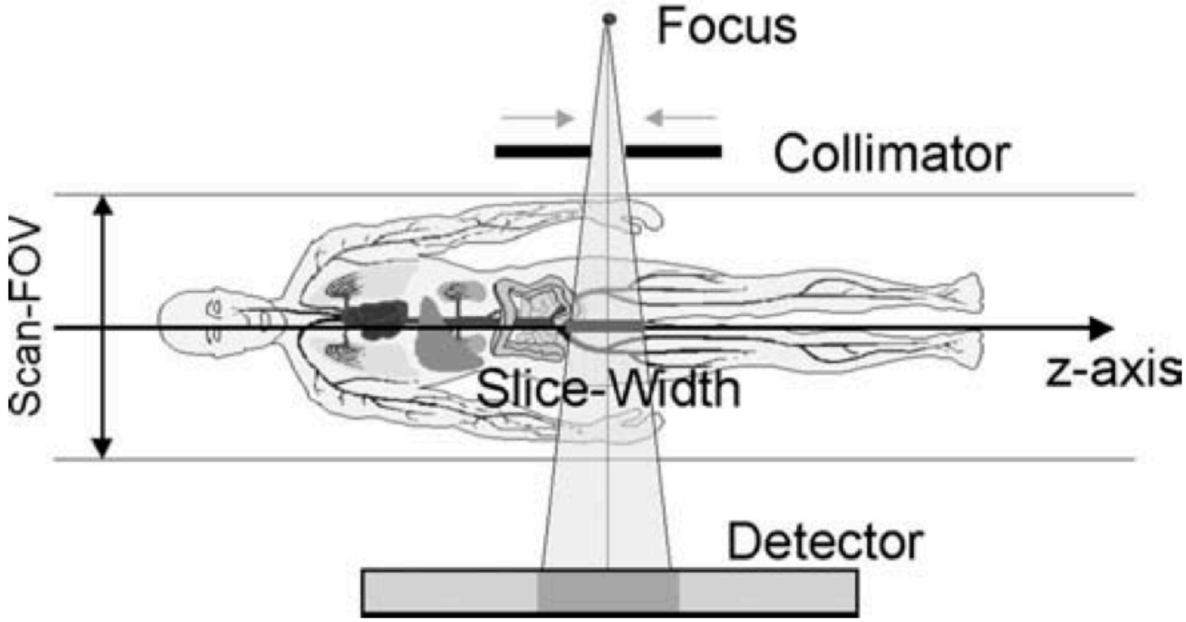
4 detector rows

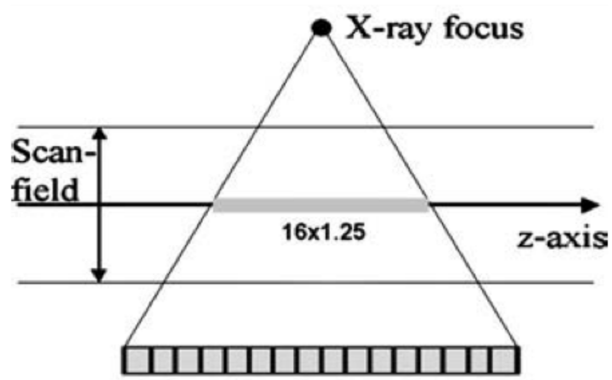
Single versus Multi-row Detectors

- Can image multiple planes at once



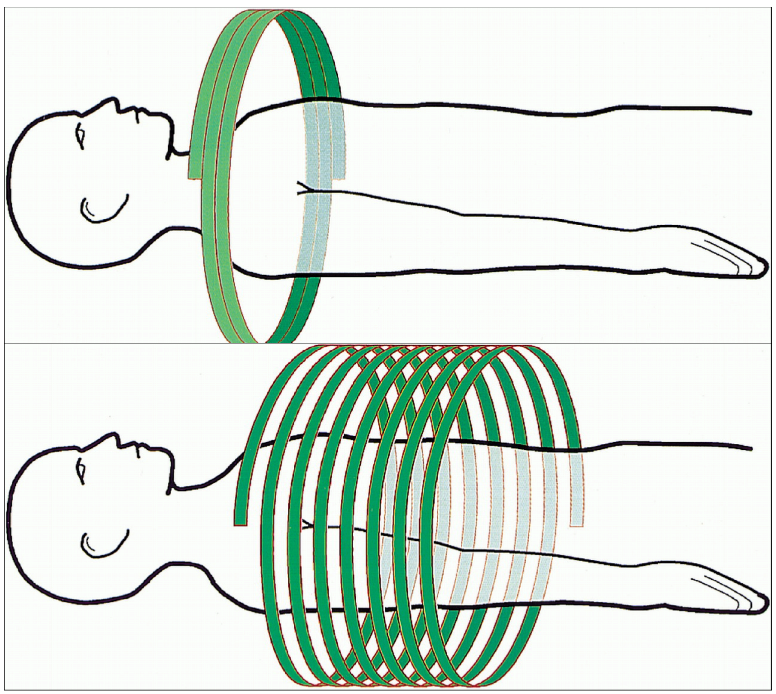
Multi-row Detectors



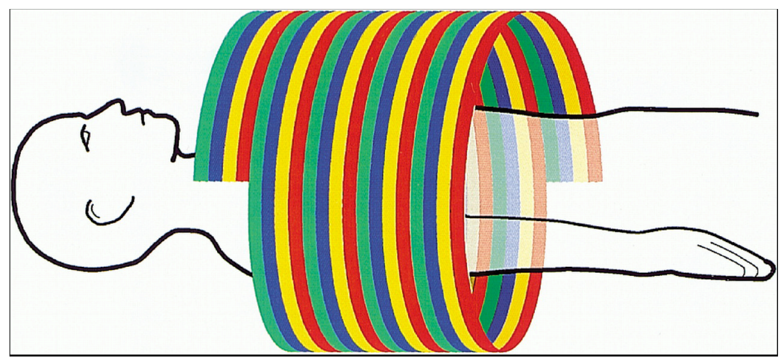


Helical Multi-Detector CT (MDCT)

- 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



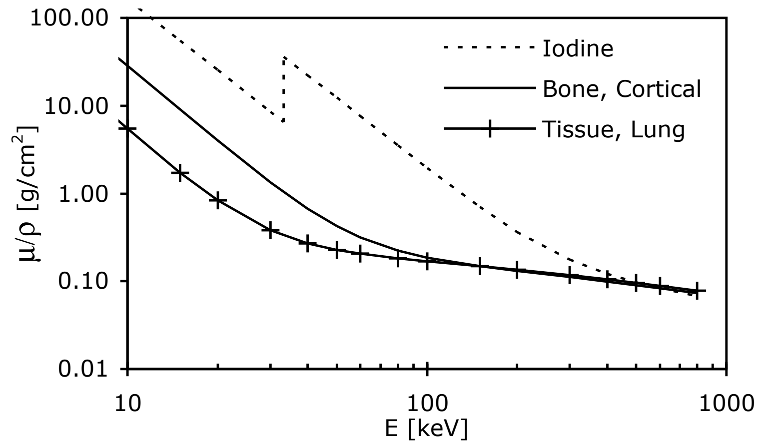
1 detector row: pitch 1 and 2



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

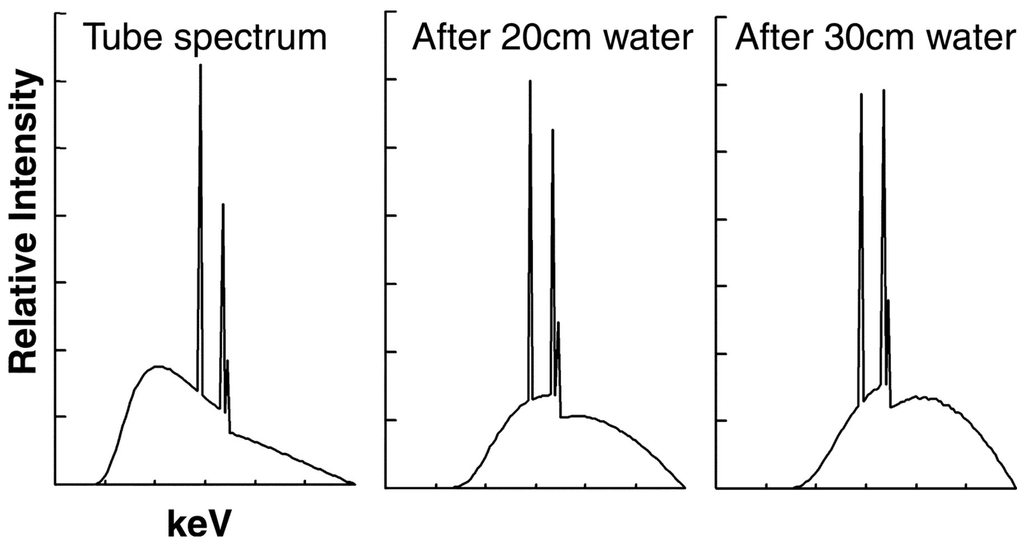
Typical Background Radiation - 3 mSv per year

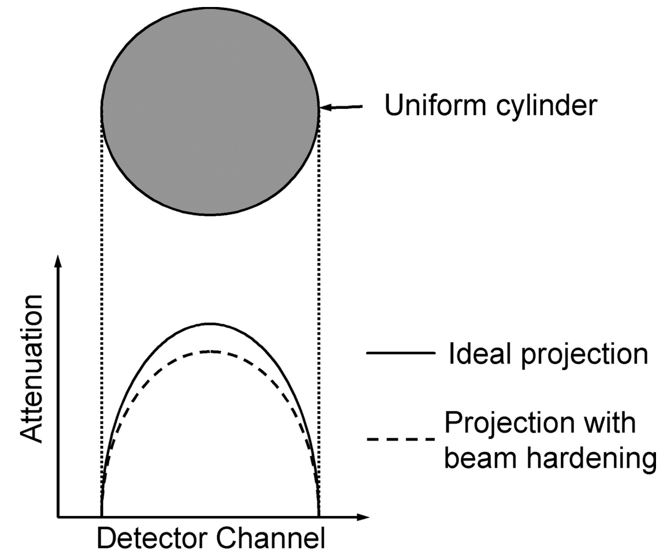
Types of CT Artifacts

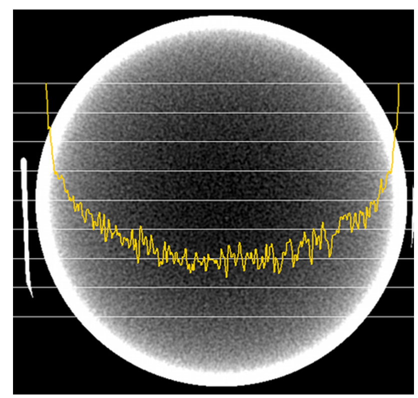
- Physics based
  - beam-hardening
  - partial volume effects
  - photon starvation
  - scatter
  - undersampling
- Scanner based
  - center-of-rotation
  - tube spitting
  - helical interpolation
  - cone-beam reconstruction
- Patient based
  - metallic or dense implants
  - motion
  - truncation

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







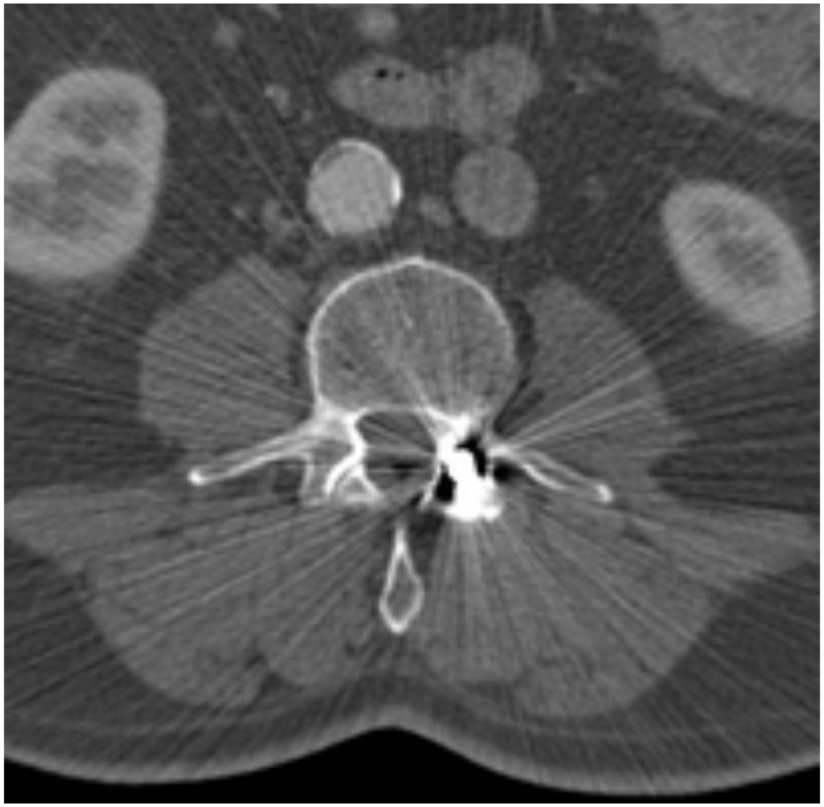
CT *image* profiles across the centre of a uniform water phantom without beam hardening correction

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