Brief Introduction to CT and PET Imaging

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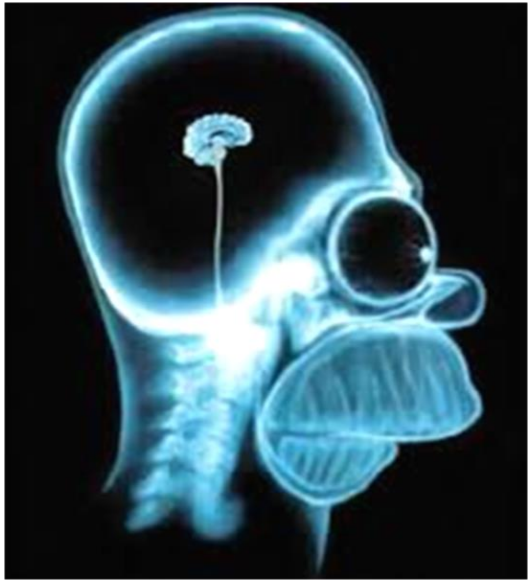
IBIC
INTEGRATED BRAIN IMAGING CENTER

Special Thanks to Paul Kinahan and Adam Alessio for sharing slides

Anatomical Imaging vs Functional Imaging

Anatomical Imaging

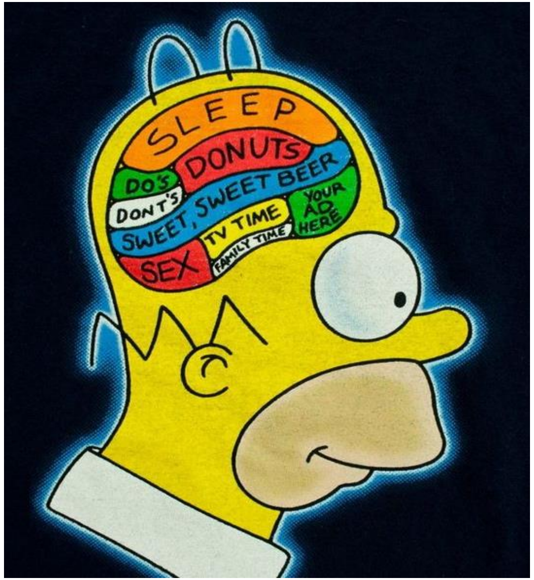
Computed Tomography (CT)



Physiological Information is **Interpreted**

Functional Imaging

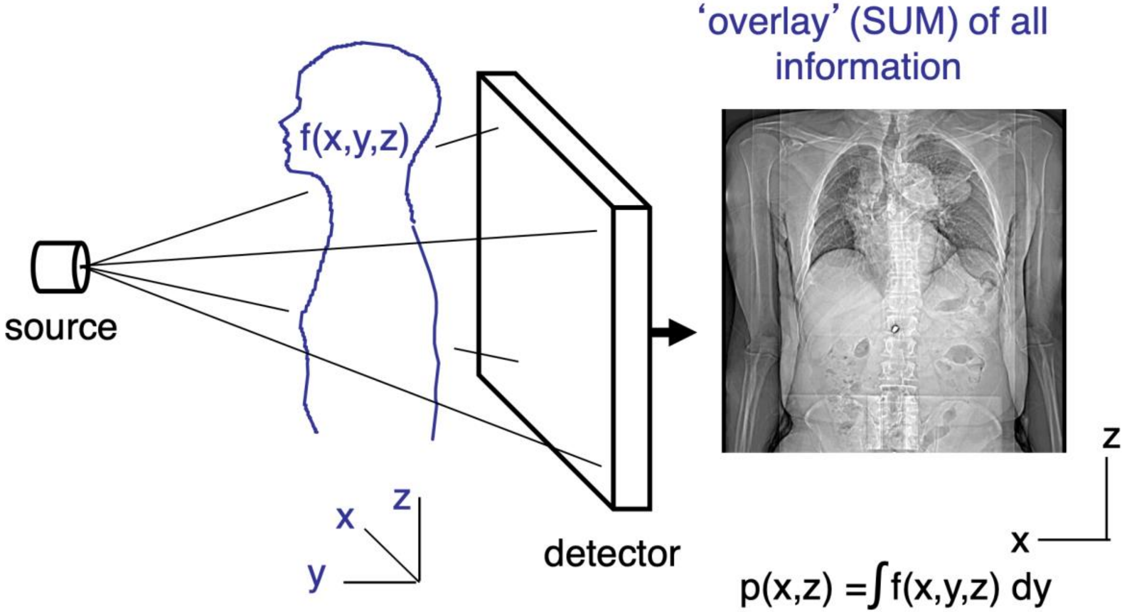
Positron Emission Tomography (PET)



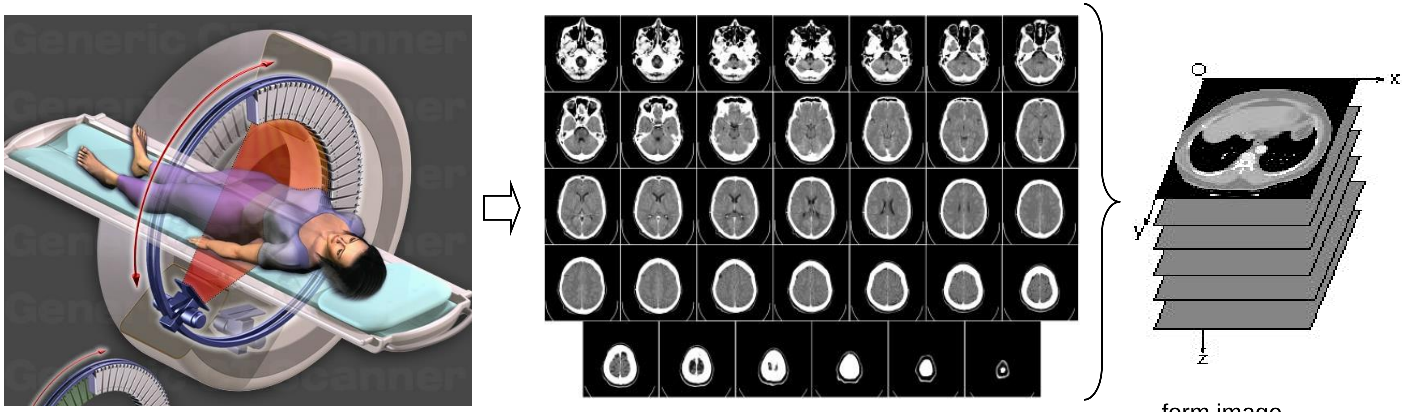
Functional Information is **Measured**

Projection Imaging

- Single 2D image is created from 3D body; “Shadow” of the body in one direction



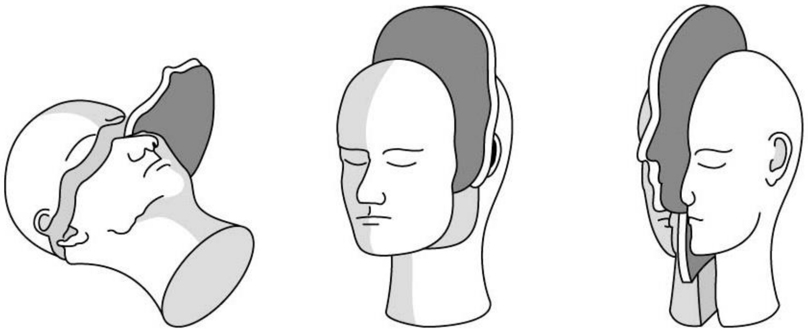
Tomographic Imaging



tomographic acquisition

reconstruction of multiple images

form image volume



Tomo + Graphy
=
Slice + Picture

transaxial or axial view

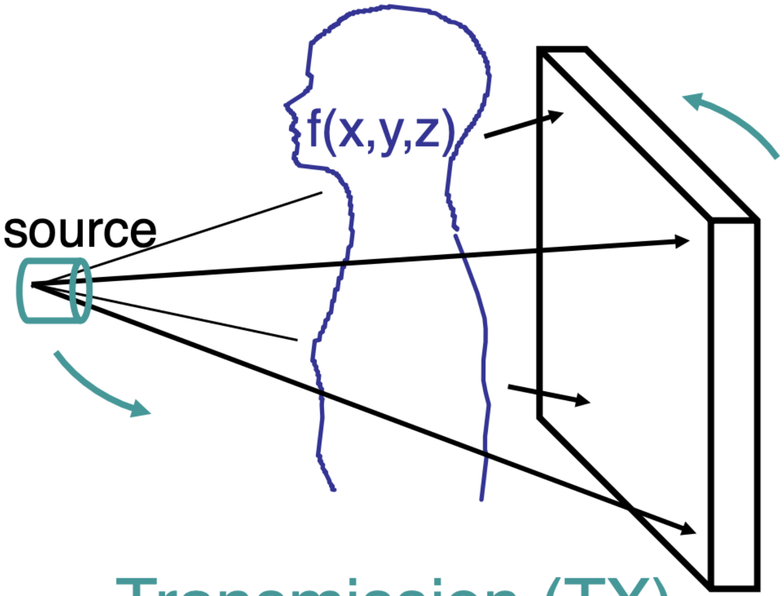
coronal view

sagittal view

True Cross-sectional Image

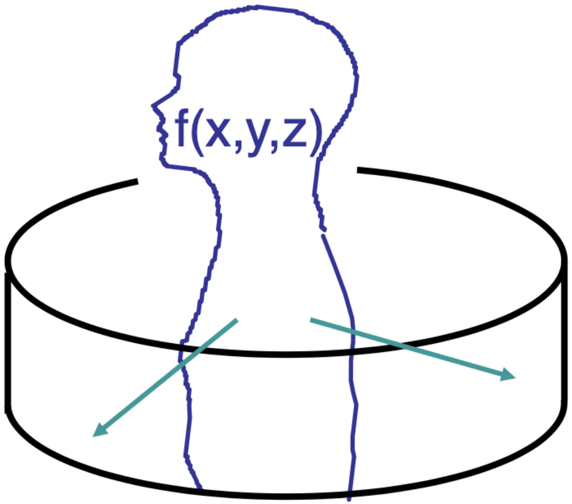
3D image from 3D body

Types of Tomographic imaging systems



Transmission (TX)

CT



Emission (EM)

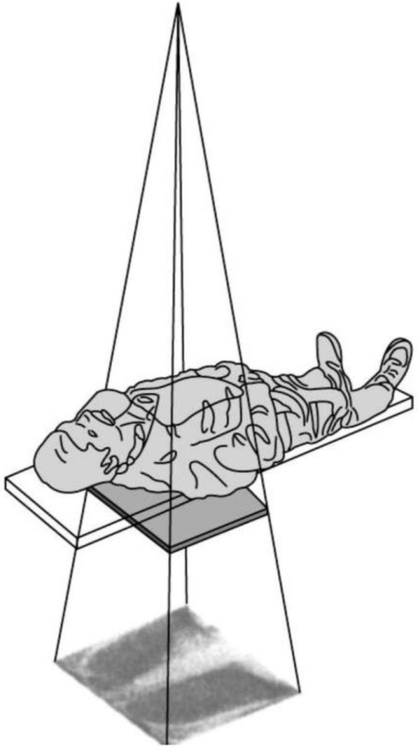
PET

Same mathematics of tomography

In the beginning, there was x-ray …

- **Contrast:** density of X-ray absorption

First x-ray image was captured in 1895 by Rontgen



The Nobel Prize in Physiology or Medicine 1979
Allan M. Cormack, Godfrey N. Hounsfield

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The Nobel Prize in Physiology or Medicine 1979

Allan M. Cormack
Prize share: 1/2

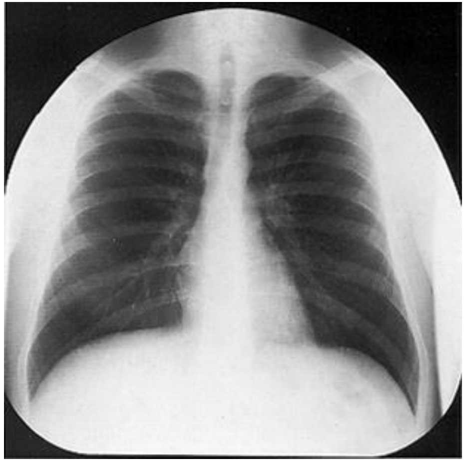
Godfrey N. Hounsfield
Prize share: 1/2

- First CT System developed at EMI Central Research Laboratories, 1967-1971
- First CT Scan of patient in 1971

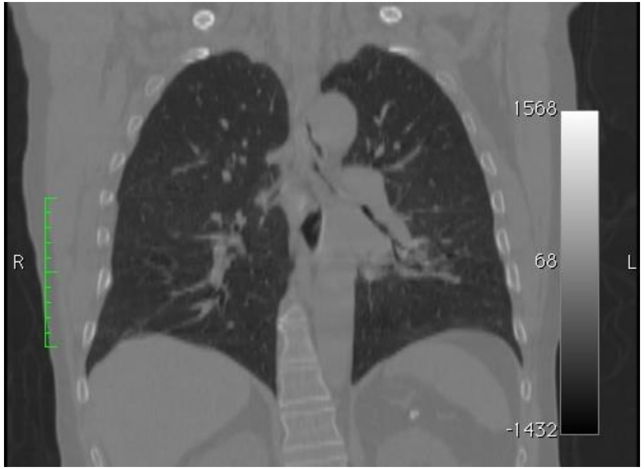
The Nobel Prize in Physiology or Medicine 1979 was awarded jointly to Allan M. Cormack and Godfrey N. Hounsfield *"for the development of computer assisted tomography"*

X-ray vs Computed Tomography (CT) Imaging

- Hounsfield's insight was that by imaging all the way around a patient we should have enough information to form a cross-sectional image
- Radiographs typically have higher resolution but much lower contrast and no depth information (i.e. in CT section below we can see lung structure)

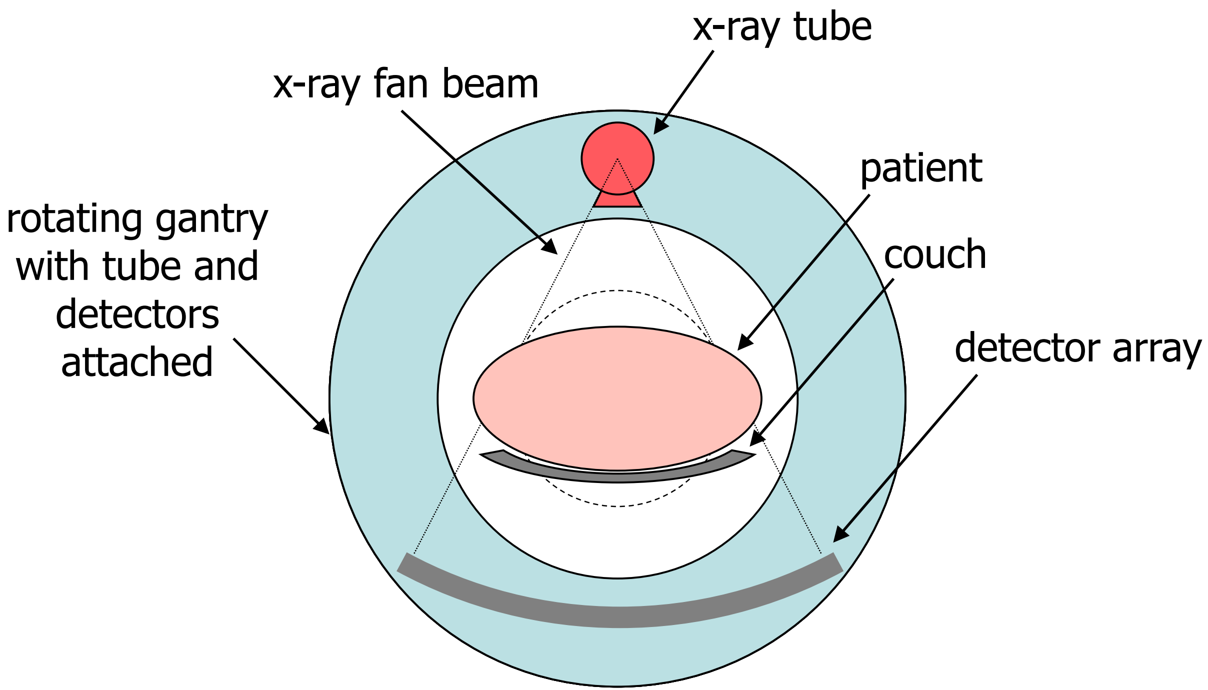


Chest radiograph



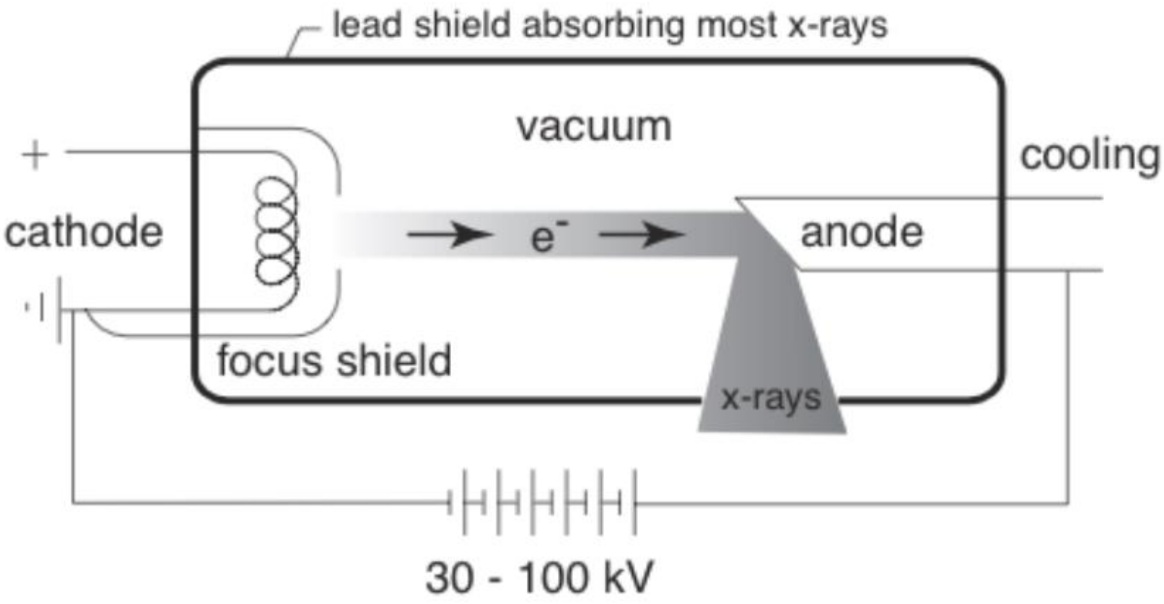
Coronal *section* of a 3D CT image volume

CT Scanner Components



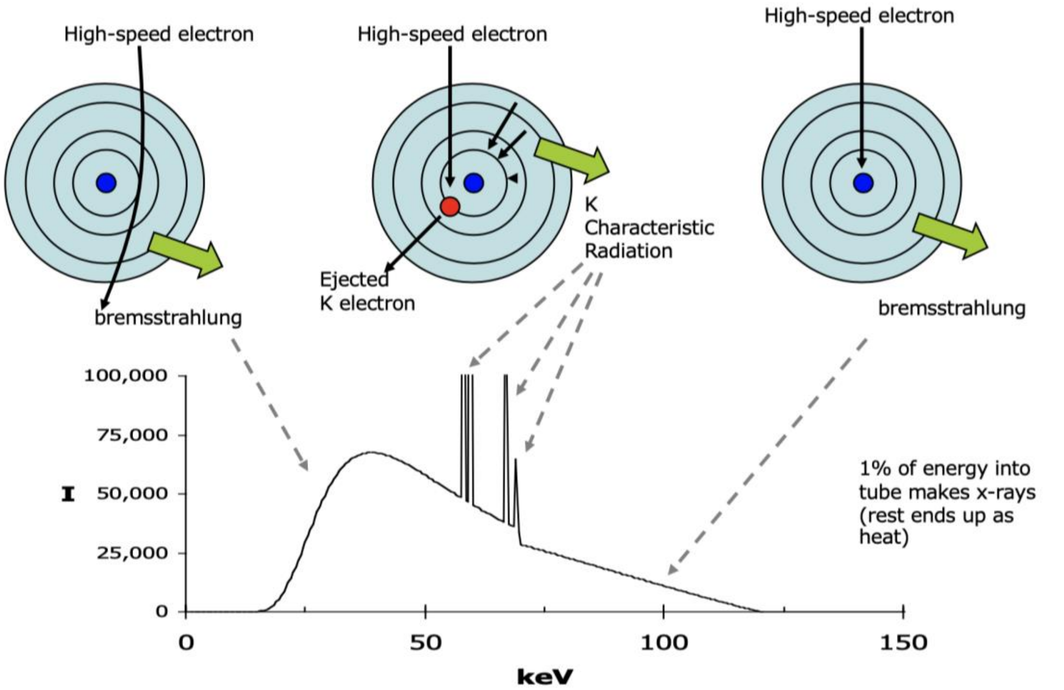
- Data acquisition in CT involves making transmission measurements through the object at angles around the object.
- A typical scanner acquires 1,000 projections with a fan-beam angle of 30 to 60 degrees incident upon 500 to 1000 detectors and does this in <1 second.

CT components: X-ray Tube



- In evacuated tube, heated cathode current releases electrons
- Electrons are accelerated to anode by voltage
- Amount of x-ray photons = Cathode current × time [mAs]
- Energy of the x-rays controlled by voltage between anode and cathode [kV]

CT components: X-ray Production

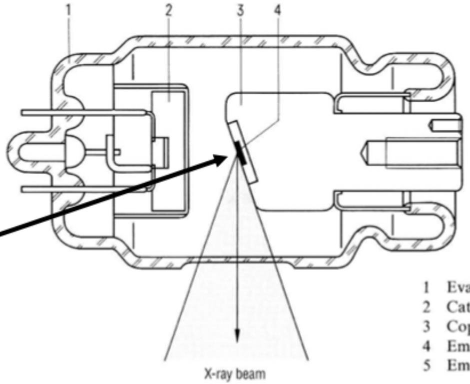


- Accelerated electron interacts with nuclei in anode
- Electrons release their energy as *bremsstrahlung* and characteristic radiation
- *bremsstrahlung* (braking radiation) is radiation produced due to the deceleration of electrons

CT components: X-ray Source Tubes

- Stationary anode tube

(low power version suitable for radiographs)



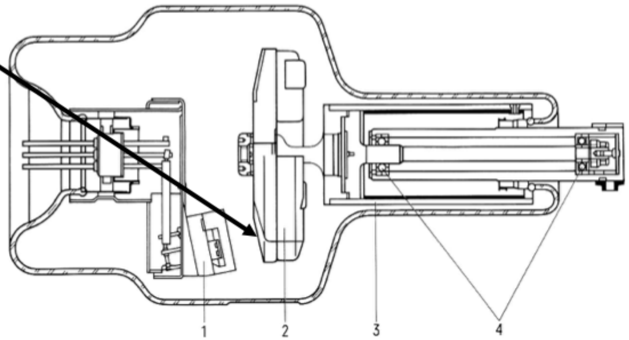
1 Evacuated glass envelope
2 Cathode
3 Copper anode
4 Embedded tungsten target
5 Emerging radiation

Fig. 7.1 Cross-section of the SR 70/7 stationary anode tube (HELIODENT)

focal spot

- Rotating anode tube

(dissipates heat for higher continuous beam currents needed for CT)

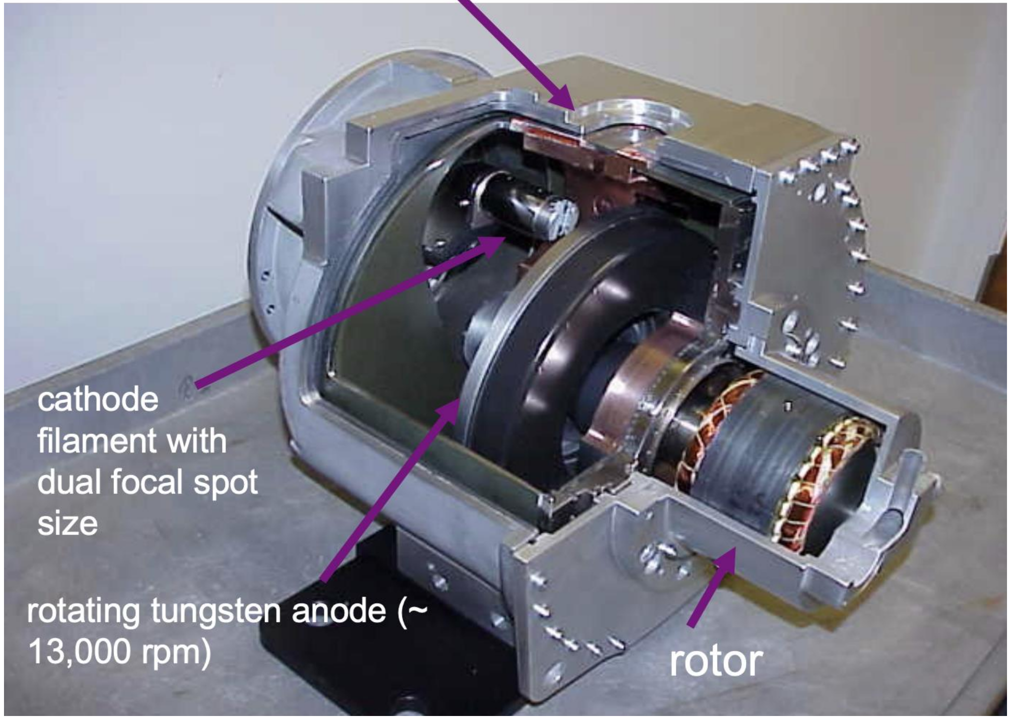


1 Dual focus cathode
2 Graphite-RTM compound anode
3 Rotor
4 Ball bearings

Fig. 7.17 Cross-section through a rotating anode x-ray tube

Modern X-ray Tube (GE Performix)

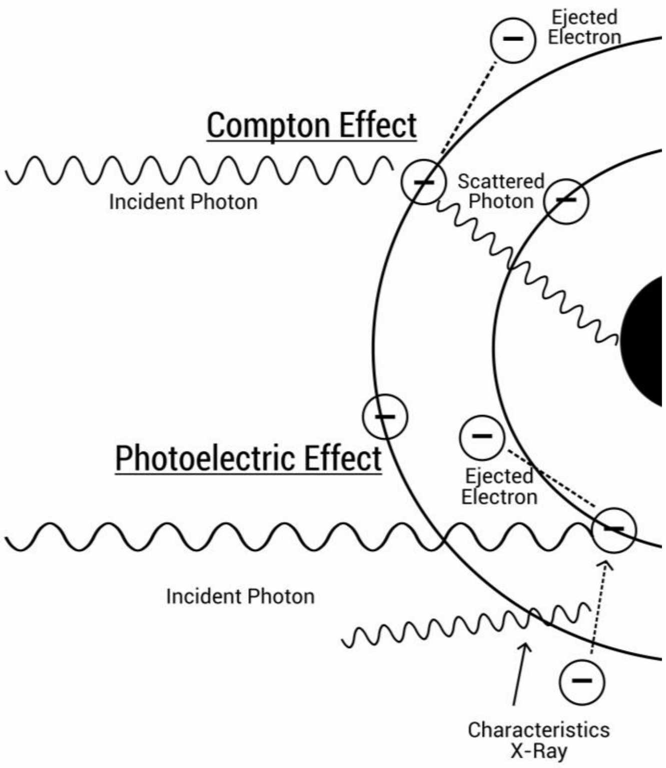
metal vacuum casing with thin metal foil for X-ray aperture



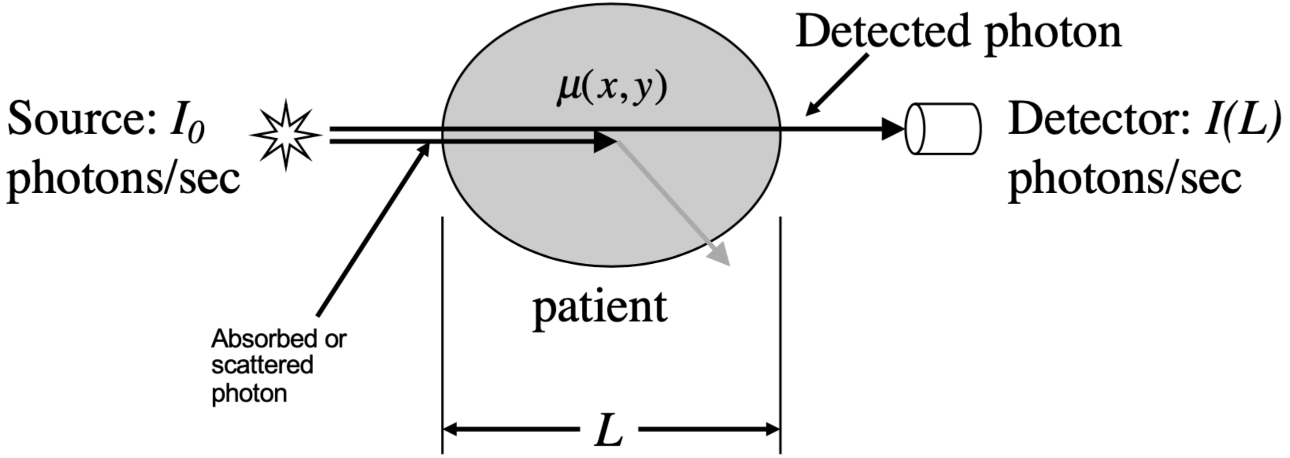
CT components: X-ray interaction with matter

Photon interacts with an outer orbital electron

Photon collides with an inner-shell electron



CT components: contrast



Narrow-beam approximation: $\frac{dI}{dx} = -\mu I \quad \rightarrow \quad I(x) = I_0 e^{-\int_0^L \mu(x.y)\,dL}$

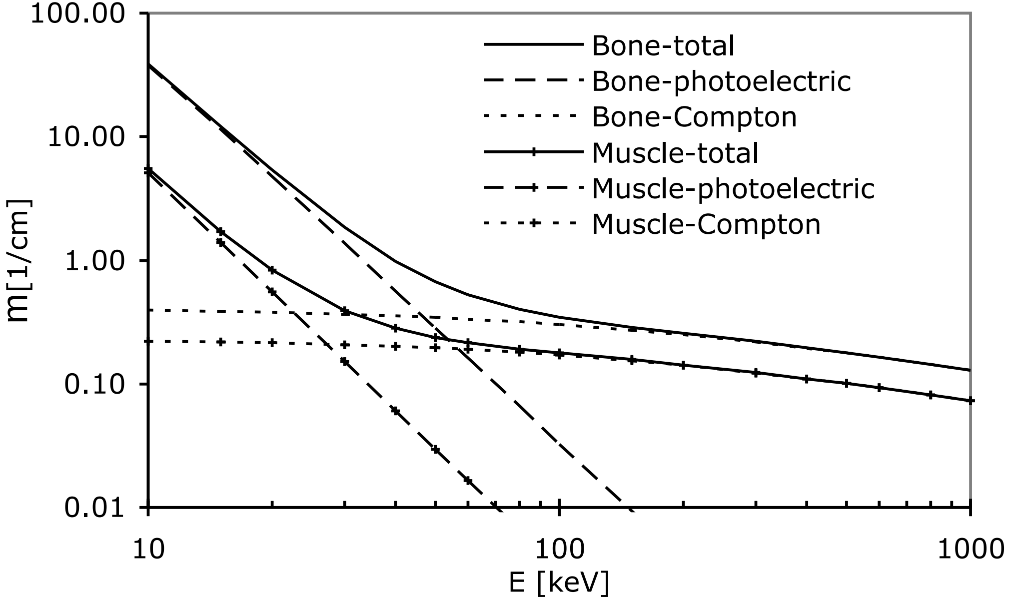
Linear attenuation coefficient (1/cm): $\mu(x,y) = \underbrace{\underset{\substack{\uparrow \\ \text{photo-electric} \\ \text{absorption}}}{\mu(x,y)_{PE}} + \underset{\substack{\uparrow \\ \text{Compton} \\ \text{scatter}}}{\mu(x,y)_{CS}}}_{\text{most important at \textasciitilde 30-1000 keV}} + \ldots$

CT components: contrast

Attenuation of photons is a function of

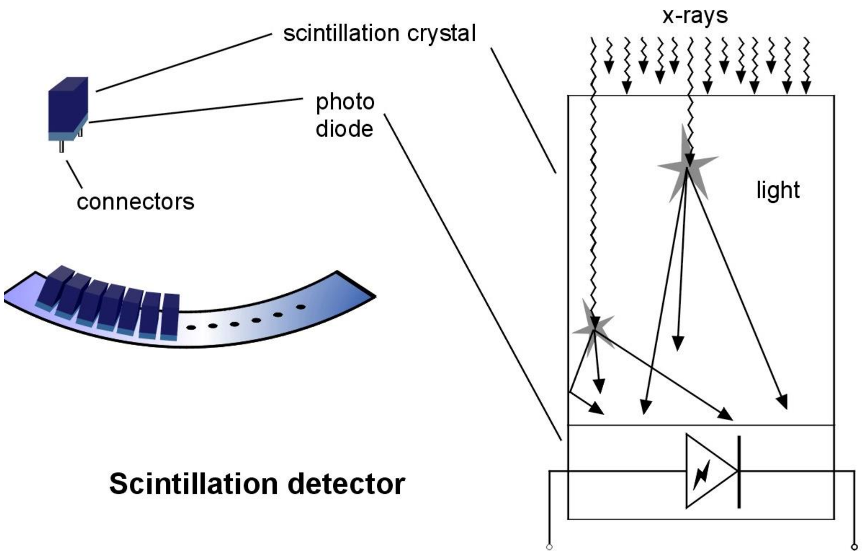
1. Energy of the photons
2. Medium (density & atomic number)

Components of linear attenuation coefficients



CT components: Detectors

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



Scintillation detector

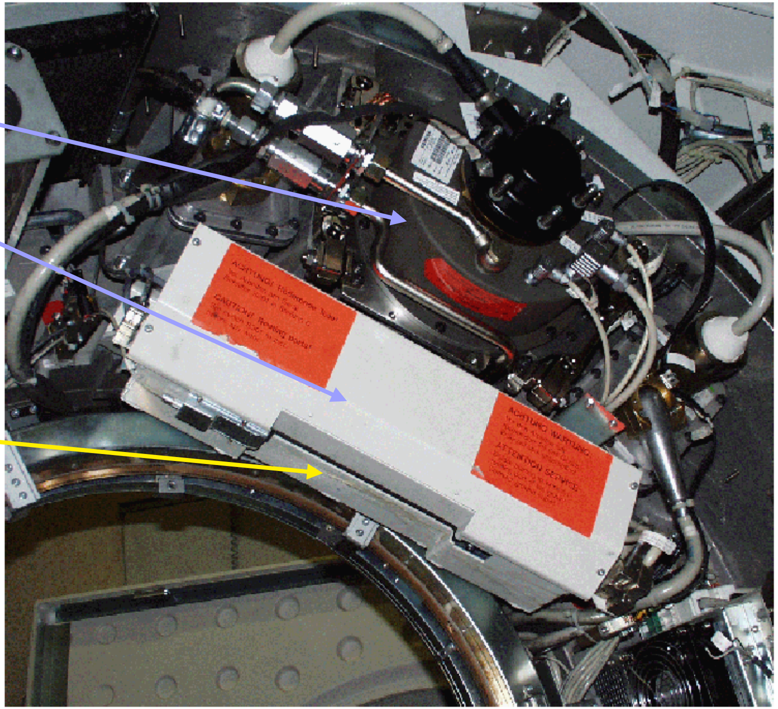
Pre-patient Collimation

- Controls patient radiation exposure

X-ray tube

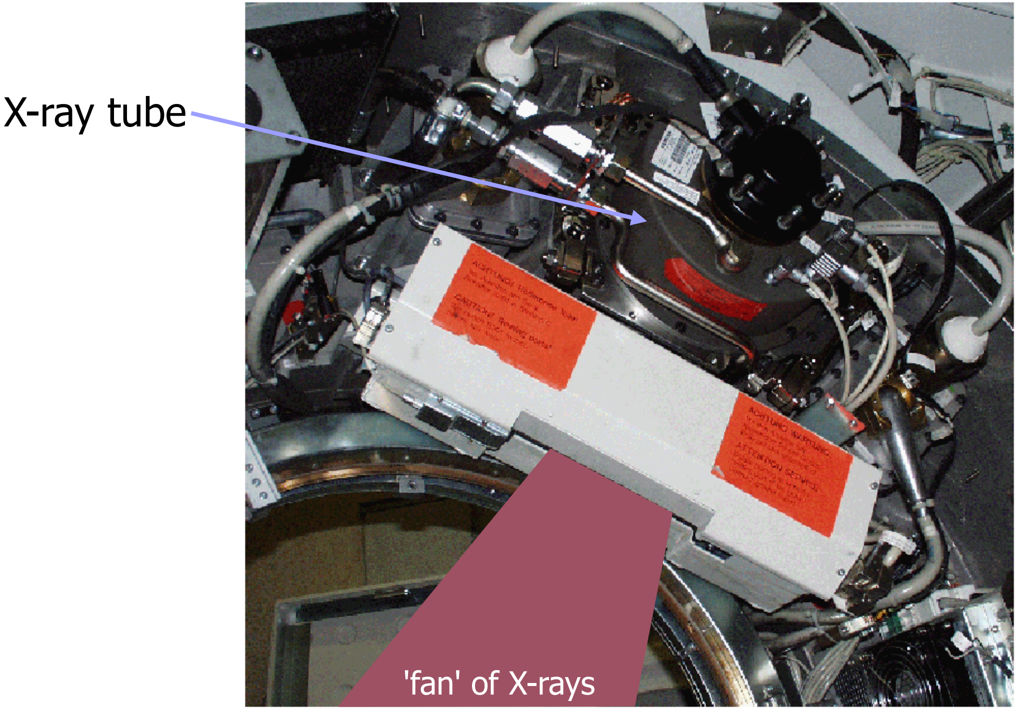
collimator
and filtration
assembly

X-ray slit

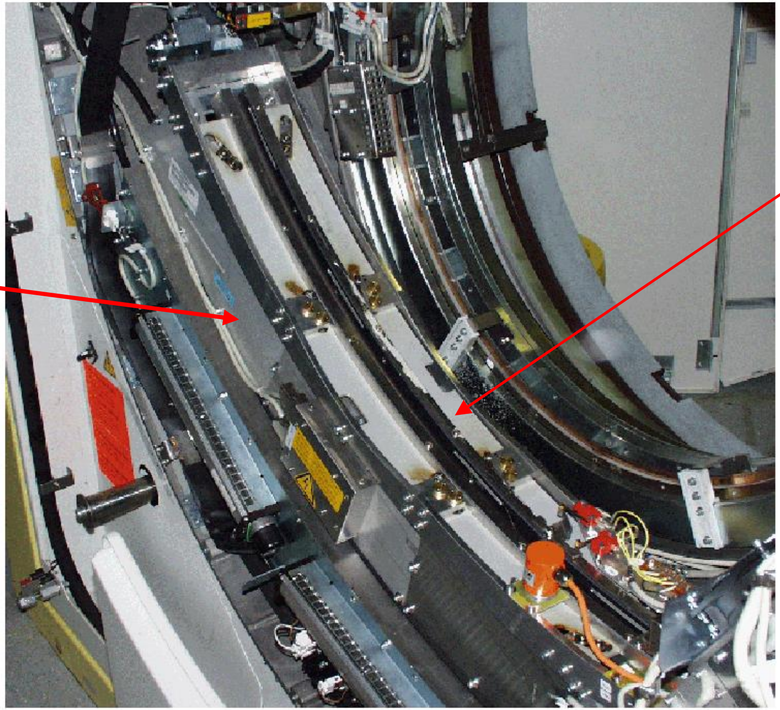


Pre-patient Collimation

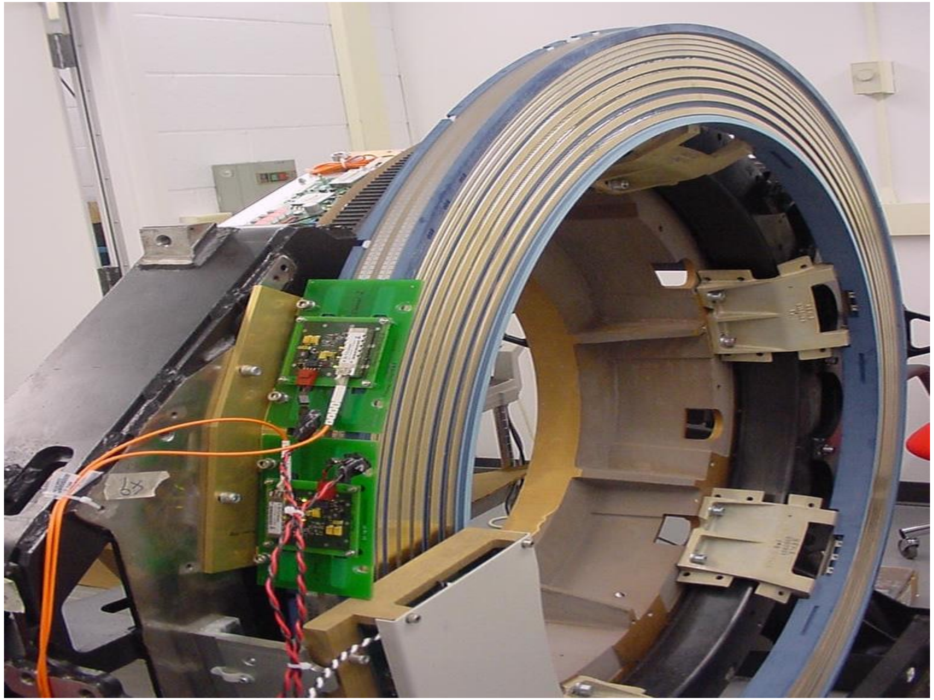
- Controls patient radiation exposure



X-ray Detector Assembly



Gantry Slip Rings



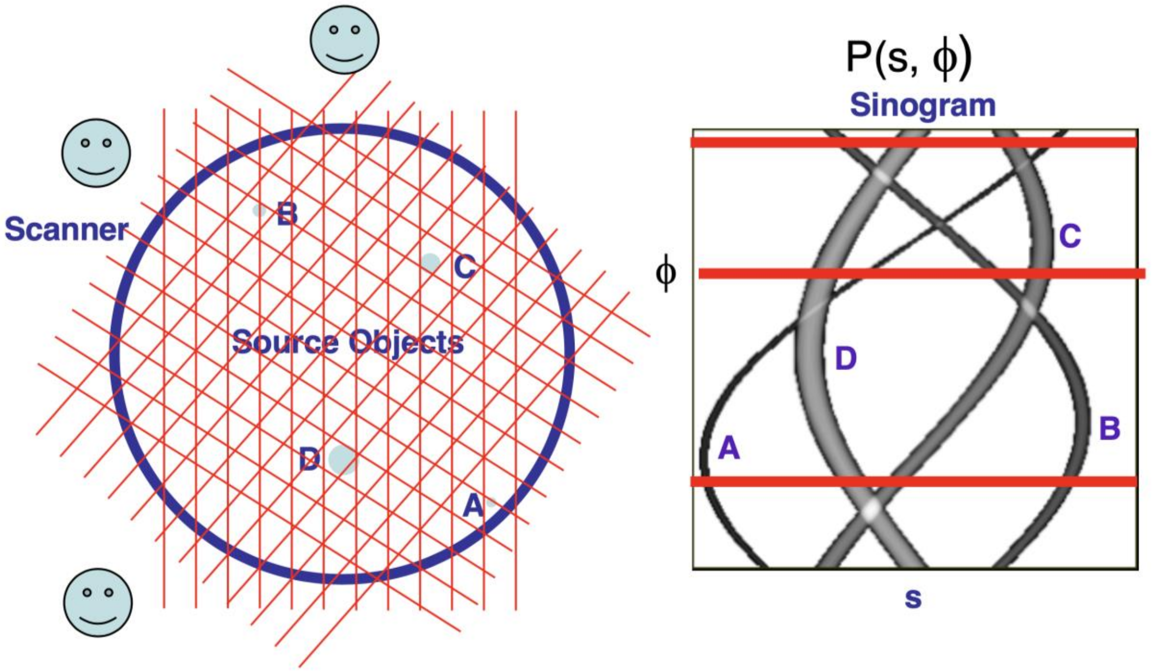
- Allows for continuous rotation

CT Scanner in operation

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

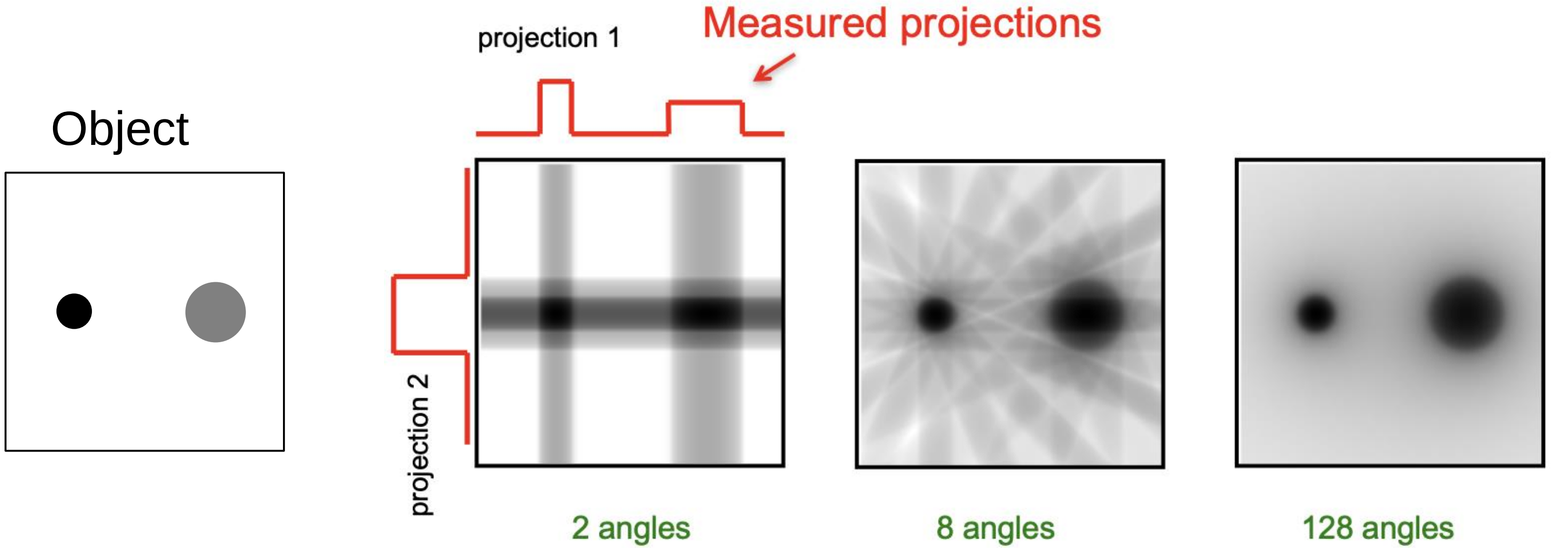
CT imaging - Summary

CT scan: Raw data



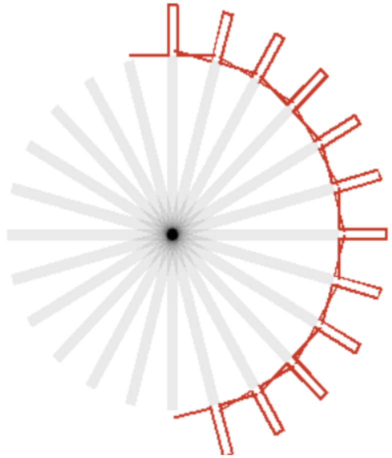
- The sinogram is P(s, ϕ) organized as a 2D histogram - Radon Transform of the object

Image reconstruction – Simple Back Projection

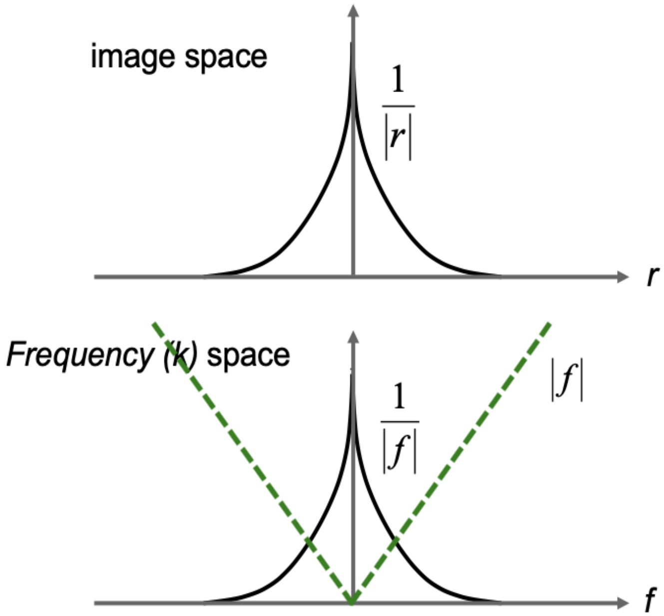


Back Projection = "smearing back" the projection across the image at the angle it was acquired

Image reconstruction – Simple Back Projection

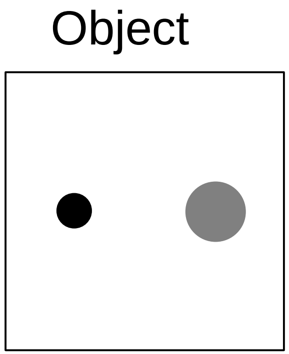


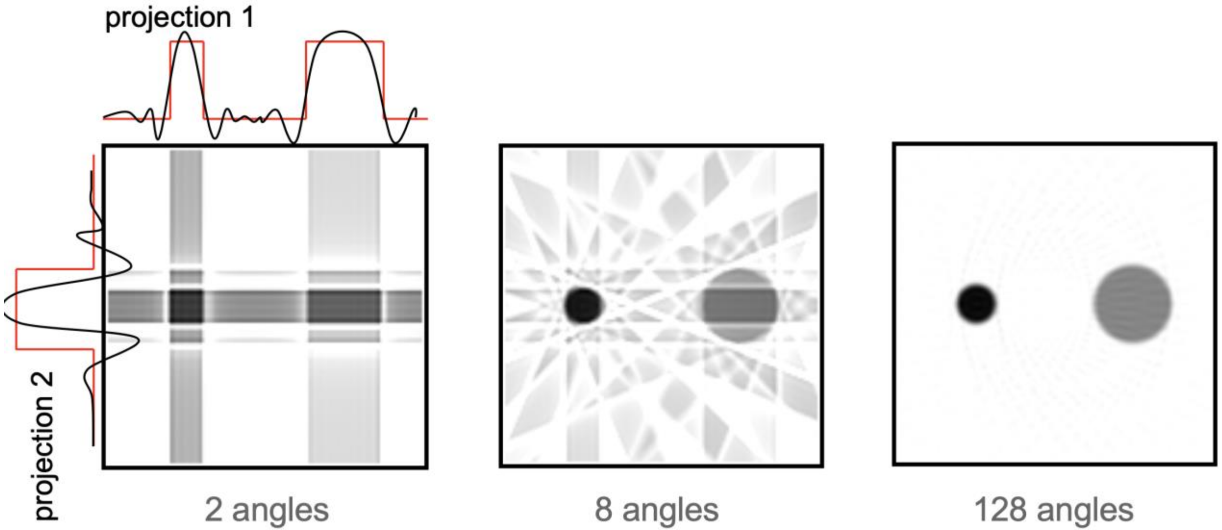
de adapted from Roger Fulton, University of Sydney



- Leads to radial (1/r) blurring in the spatial domain
- Equivalent to a 1/f function in the frequency domain
- We need to multiply 1/f by |f| to result in a perfect point source reconstruction

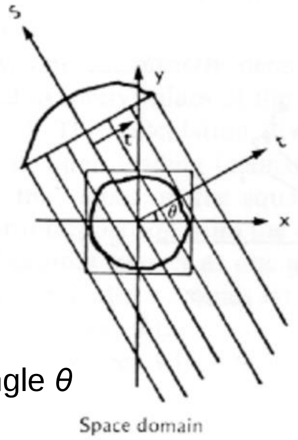
Image reconstruction – Filtered Back Projection





Filtered Back Projection = "smearing back" the *filtered* projection across the image at the angle it was acquired

Filtered Back Projection



P(t, θ)

Sinogram

Projections:

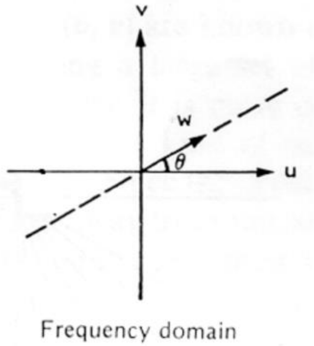
$$P_\theta(t) = \int_{(\theta,t)\,line} f(x,y)\,ds$$

<u>Projection Slice Theorem:</u> Fourier transform of a projection at angle θ gives us a line in the polar Fourier space at the same angle θ

Inverse Fourier Transform Image:

$$f(x,y) = \int_{-\infty}^{\infty}\int_{-\infty}^{\infty} F(u,v)e^{j2\pi(ux+vy)}\,du\,dv$$

We didn't sample *F(u,v)* in cartesian coordianate *u,v*
We sampled in polar coordinates *w,theta:*



$$u = w\cos\theta$$

$$v = w\sin\theta$$

$$du\,dv = w\,dw\,d\theta$$

Origin of Ramp Component

$$f(x,y) = \int_0^{2\pi}\int_0^{\infty} F(w,\theta)e^{j2\pi w(x\cos\theta + y\sin\theta)}\boxed{w}\,dw\,d\theta$$

Filtered Back Projection

$$f(x,y) = \int_0^{2\pi} \int_0^{\infty} F(w,\theta) e^{j2\pi w(x\cos\theta + y\sin\theta)} w \, dw \, d\theta$$

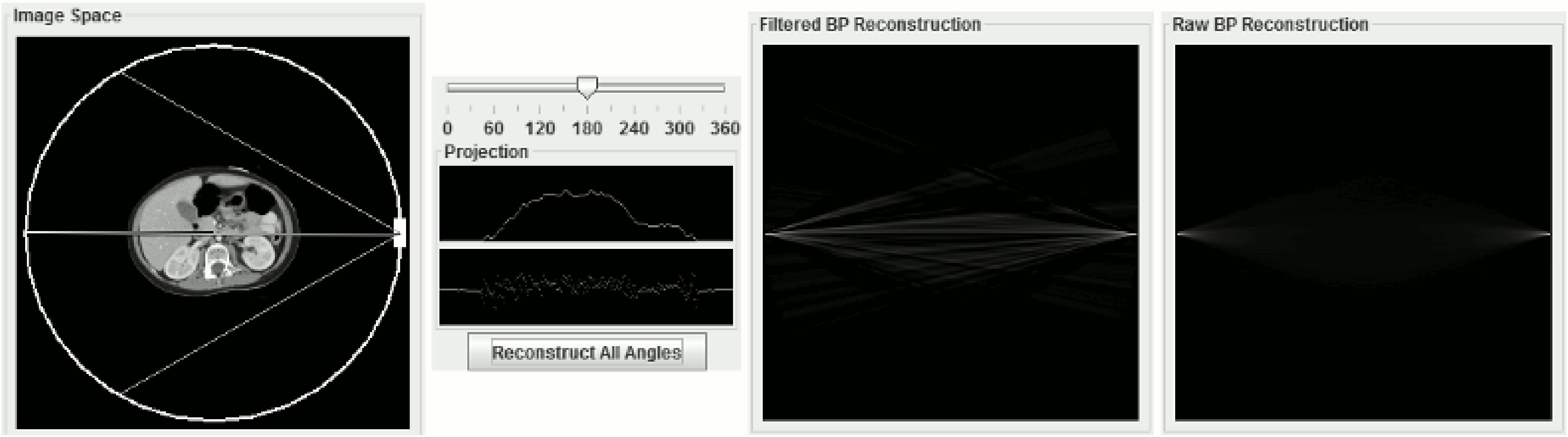
Use Symmetry Properties: $F(w,\theta+180) = F(-w,\theta)$

Substitution: $t = x\cos\theta + y\sin\theta$

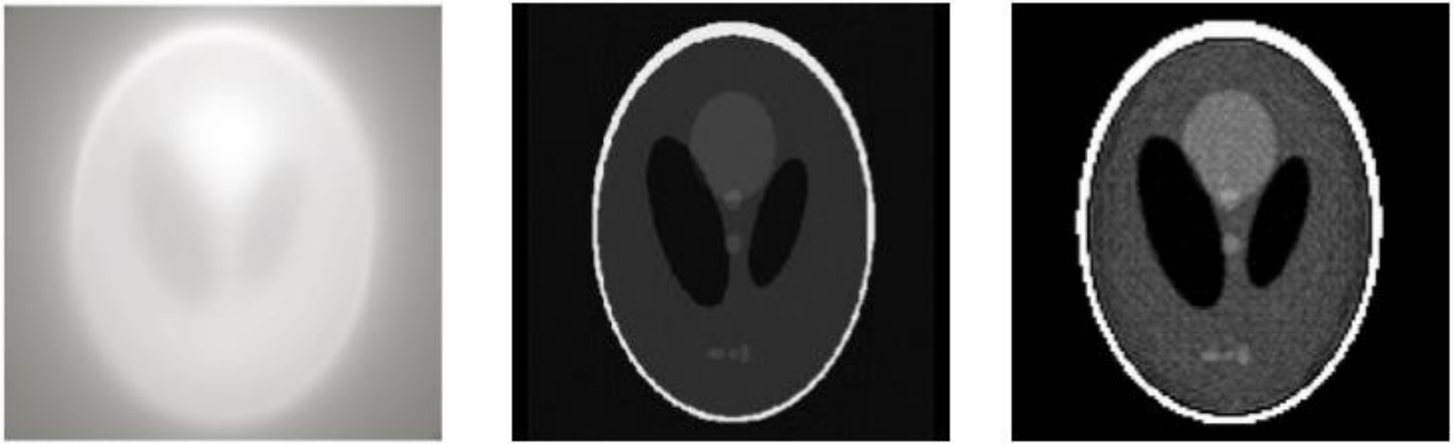
FBP Algorithm: $f(x,y) = \int_0^{\pi} \left[\int_{-\infty}^{\infty} F(w,\theta) |w| e^{j2\pi wt} \, dw \right] d\theta$

1. Take Fourier Transform of a projection at θ
2. Weight with "ramp" w
3. Take Inverse Fourier Transform at θ
4. Integrate over all θ to get $f(x,y)$

Image reconstruction – Filtered Back Projection



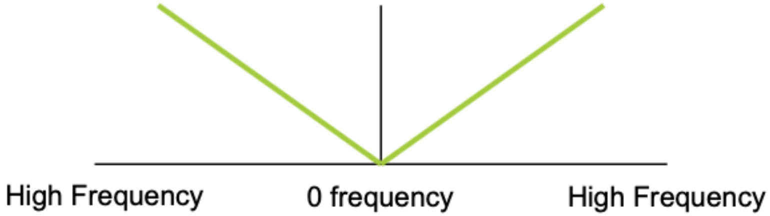
Filtered Back Projection



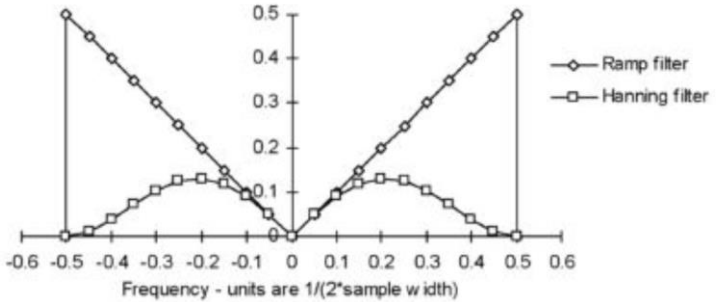
BP | FBP without noise | FBP with noise

Works well when there is no noise, but real projections contain noise...

Filtered Back Projection – Effects of Filter

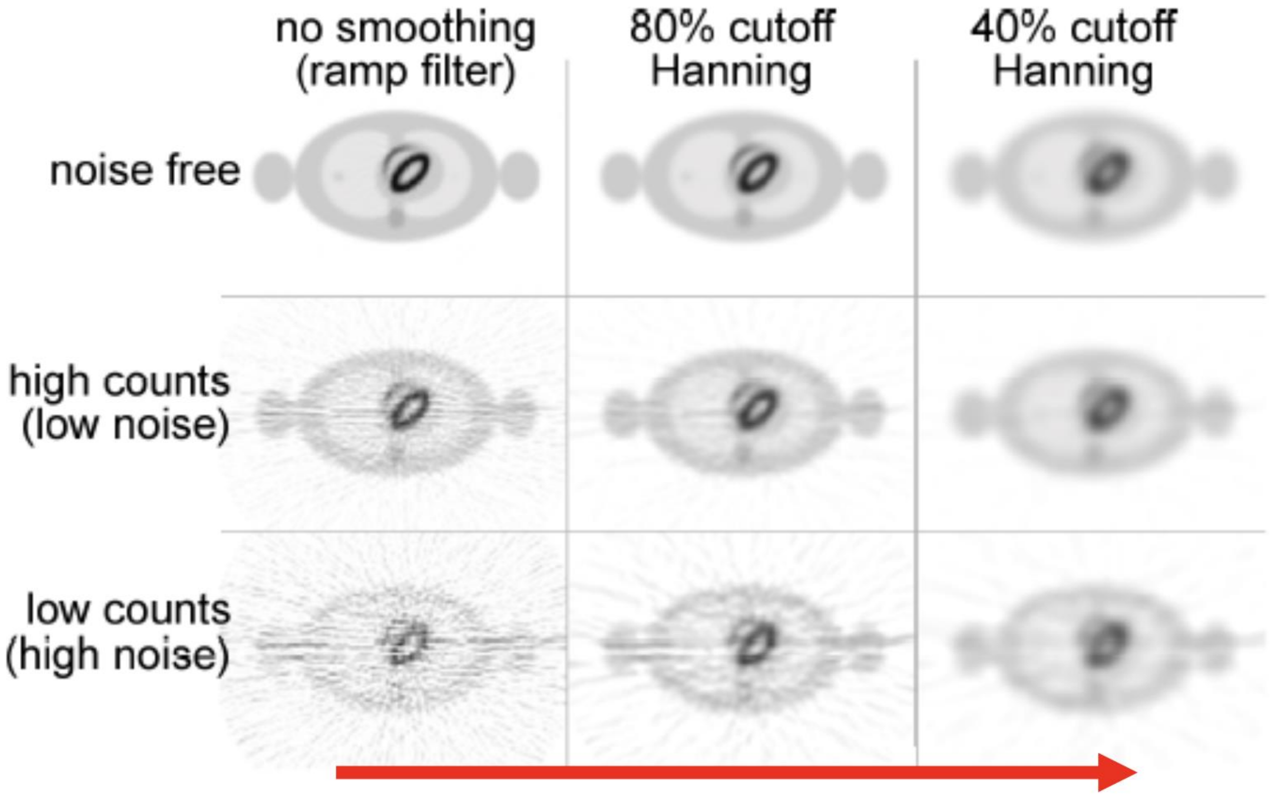


Ramp filter accentuates high frequency - Not good for noise



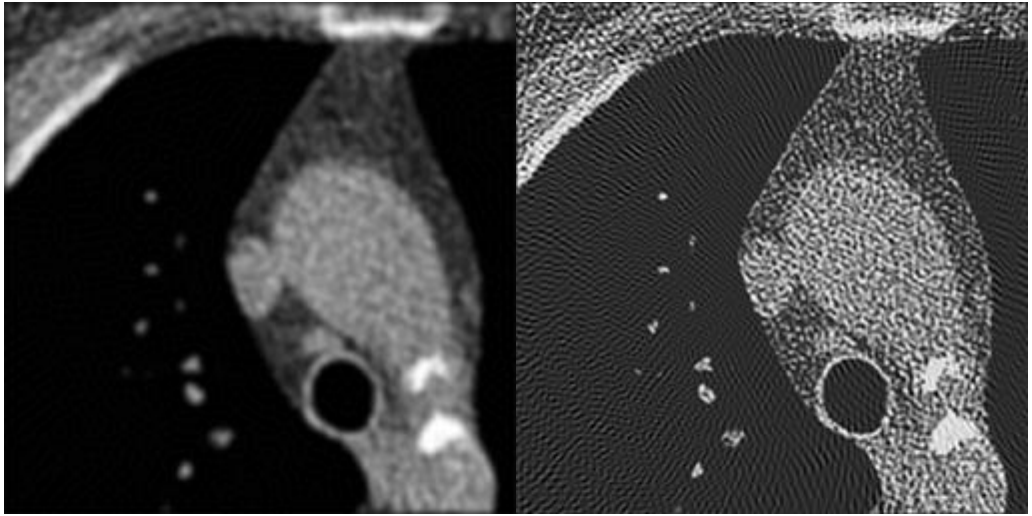
Additional filters are used to roll-off ramp to reduce noise: ex: Hanning filter

Filtered Back Projection – Effects of Filter



Reduce noise & Increase blurring

Filtered Back Projection – Effects of Filter



Soft Kernel | Sharp Kernel

We choose different kernels based on what type of image we are trying to create.

Filtered Back Projection – Effect of fan-beam angle

- In a fan-beam geometry, the angle of the fan determines how much of the object is included in the reconstructible field of view.
- A point must be included in all 180 degrees of projections in order to be reconstructed correctly.

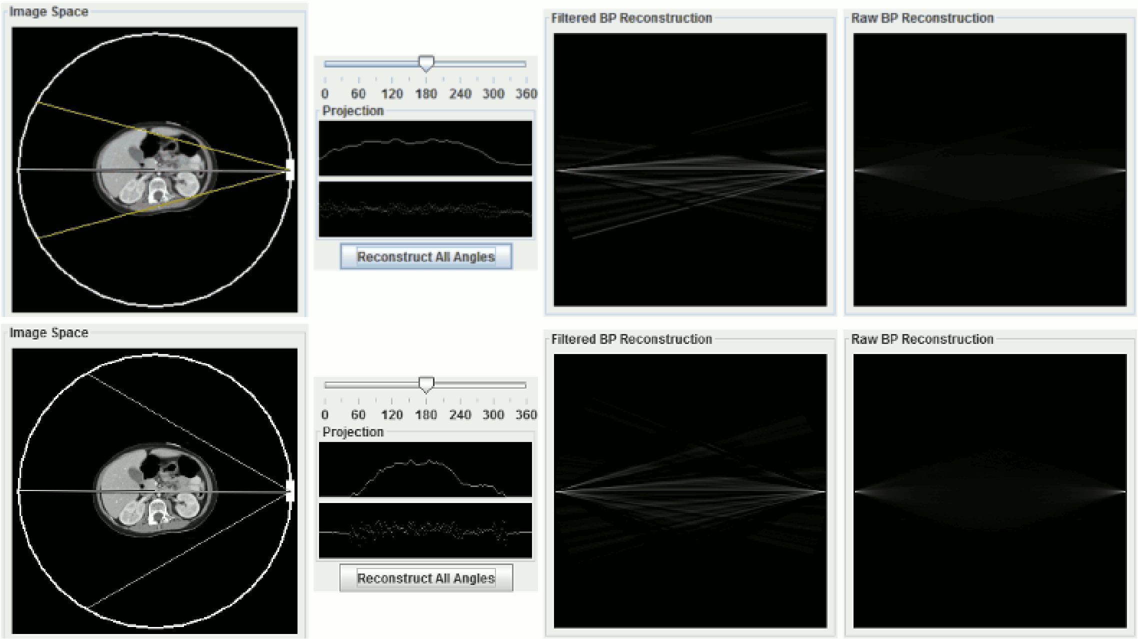
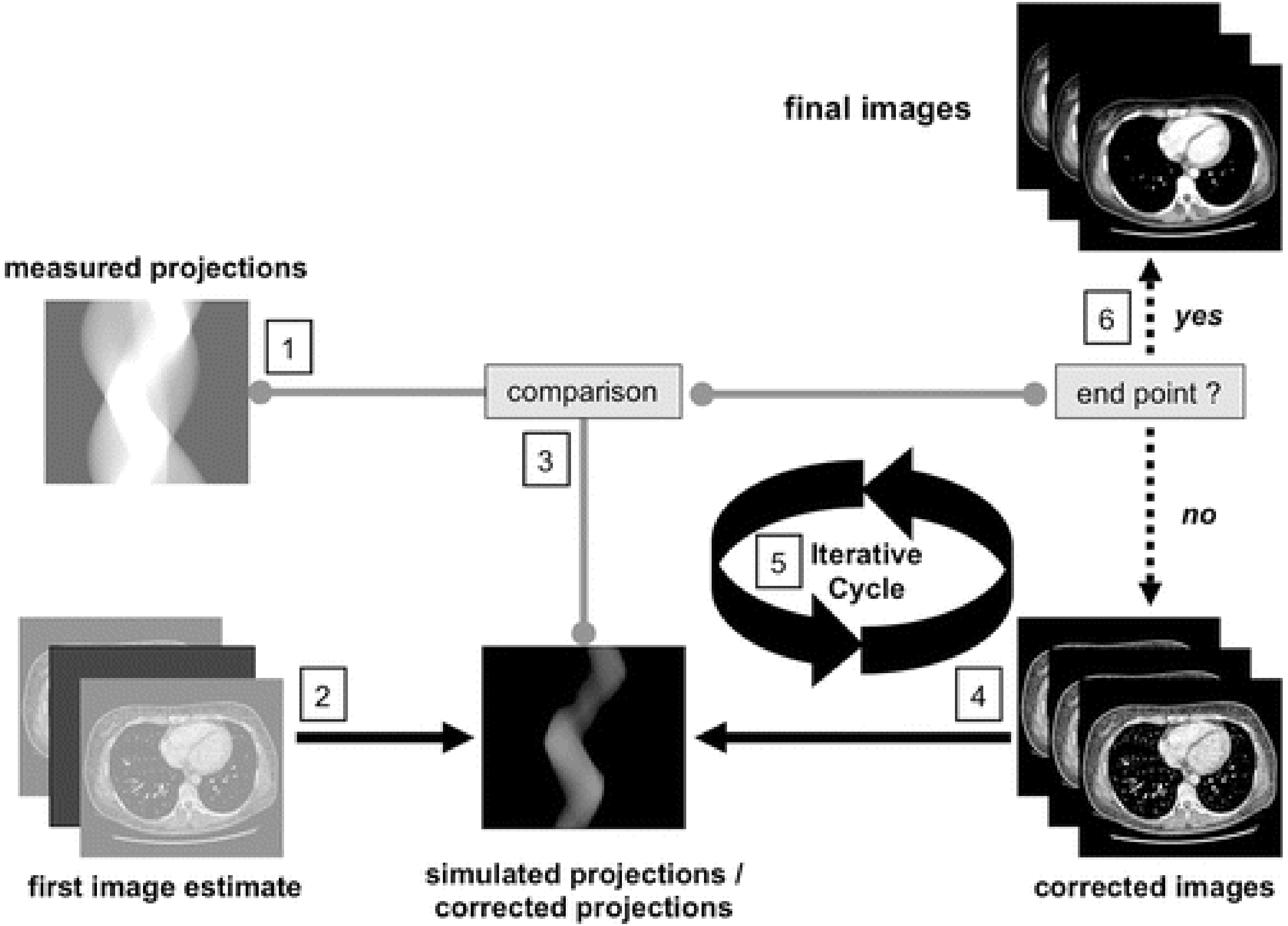


Image reconstruction – Iterative Methods

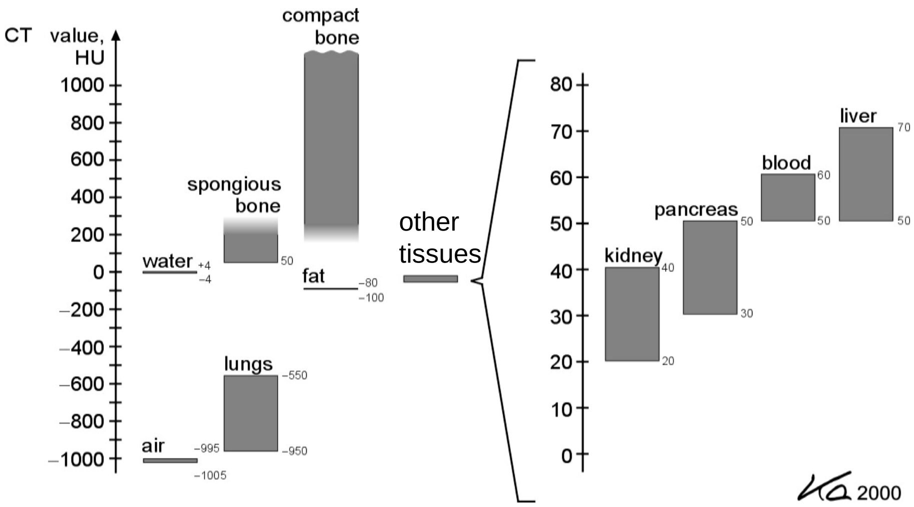


CT Number

- To provide a uniform scale, do a scale conversion where all scanners set air = -1000 and water to 0. Units are called Hounsfield units [HU] after Godfrey N. Hounsfield.

$$\text{CT number} = 1000\,\frac{\mu_{pixel} - \mu_{water}}{\mu_{water}}\,[\text{HU}]$$

- Some typical values



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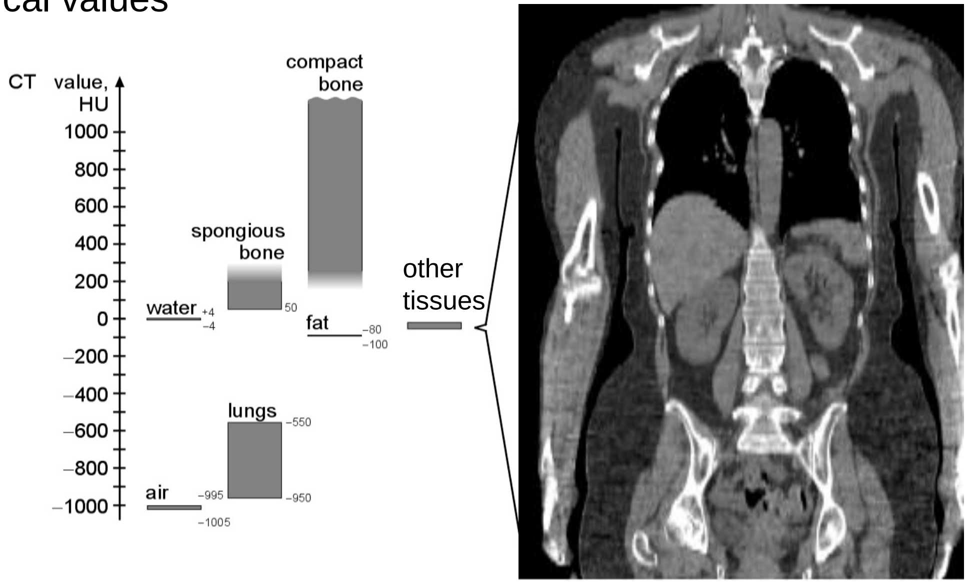
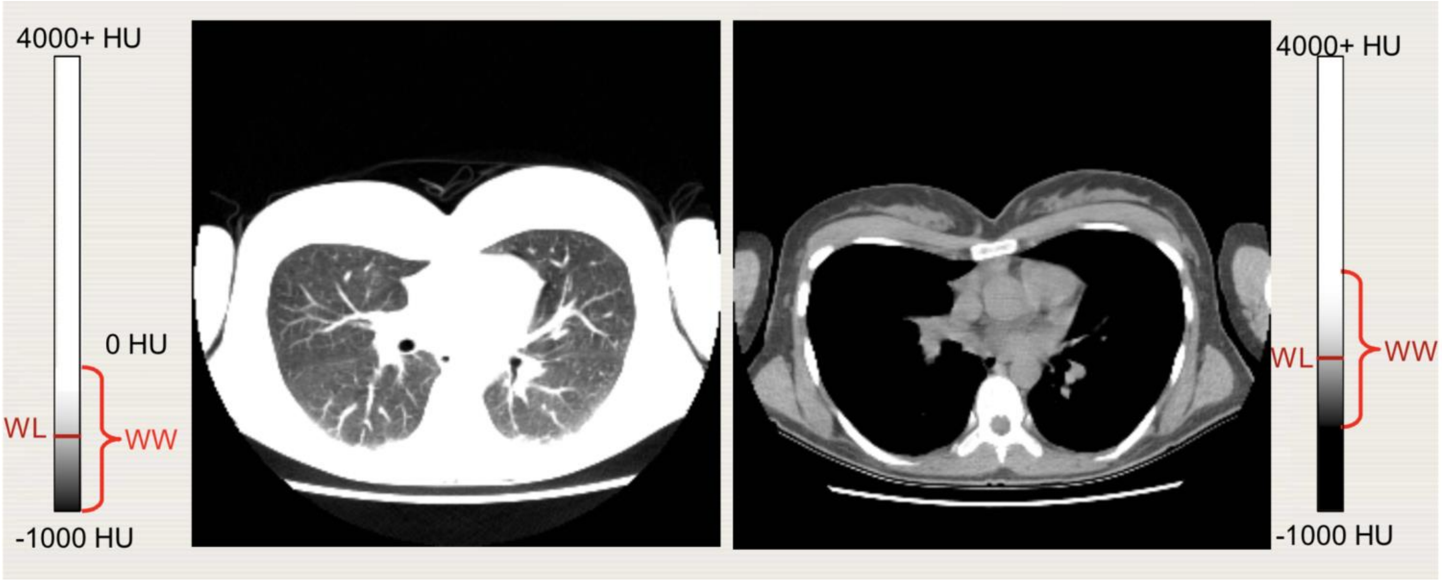


CT Number – Displaying CT Images

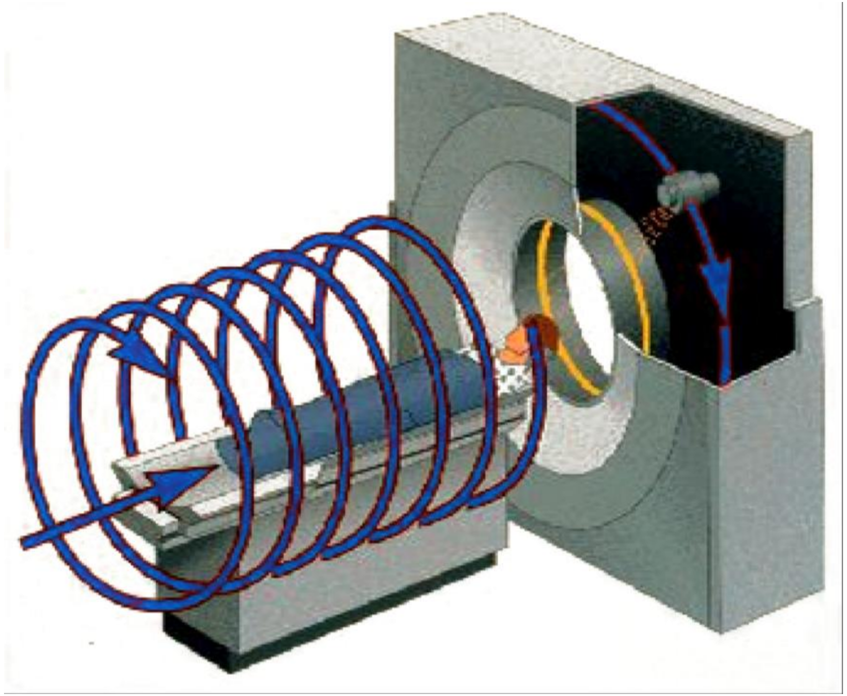


WL -593, WW 529 WL -12, WW 400

- Same image data at different WL and WW

Helical CT scanning

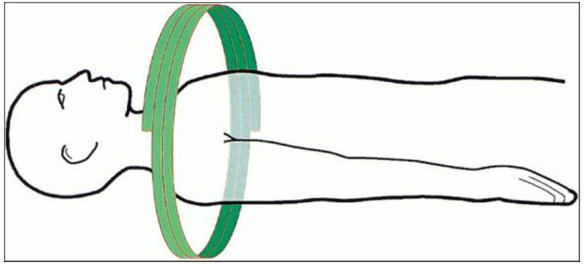
- The patient is transported continuously through gantry while data are acquired continuously during several 360-deg rotations
- The ability to rapidly cover a large volume in a single-breath hold eliminates respiratory misregistration



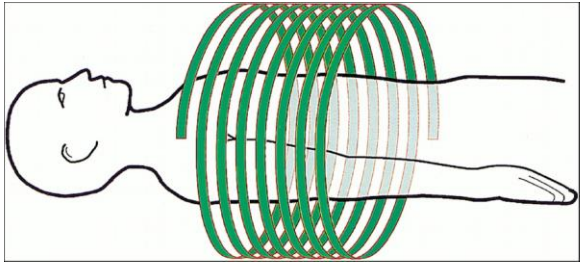
Helical CT scanning

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

slingle slice example



Pitch = 1

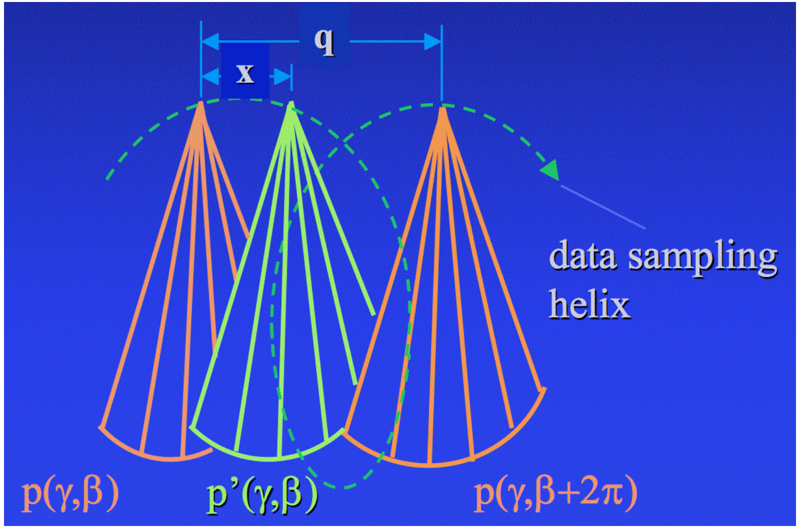


Pitch = 2

- 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).

Helical CT scanning – Image reconstruction

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

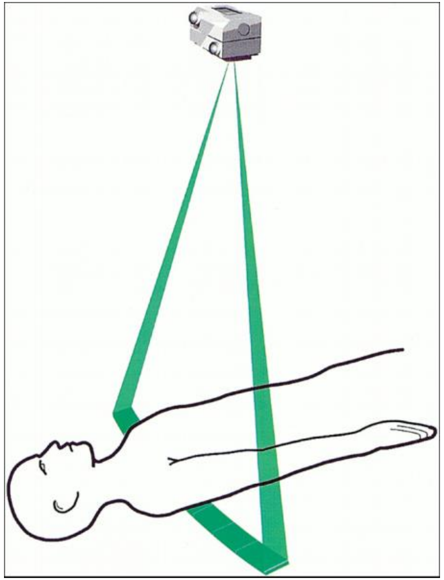


$$p'(\gamma,\beta) = w p(\gamma,\beta) + (1 - w) p(\gamma,\beta + 2\pi)$$

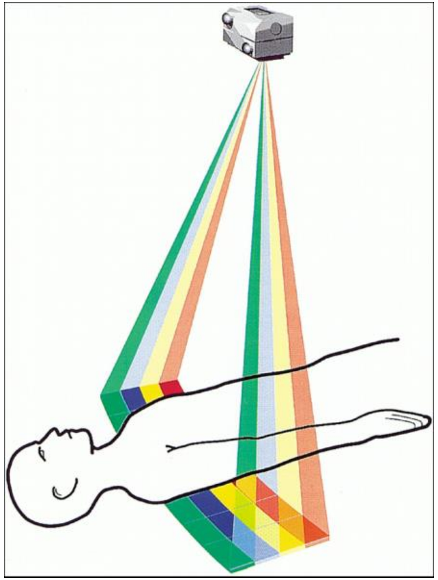
where $w = (q - x)/q$

Single vs Multi-row detectors

Can image multiple planes at once



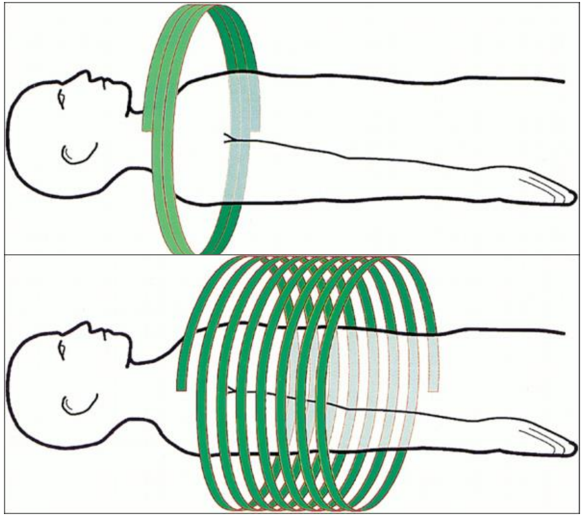
1 detector row



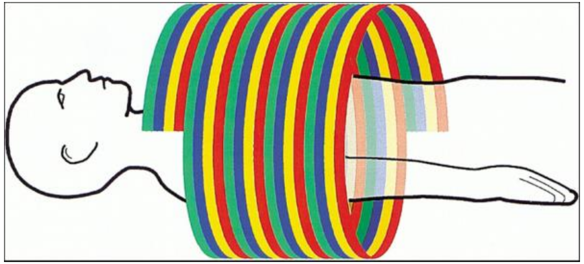
4 detector rows

Helical-multidetector CT

- 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



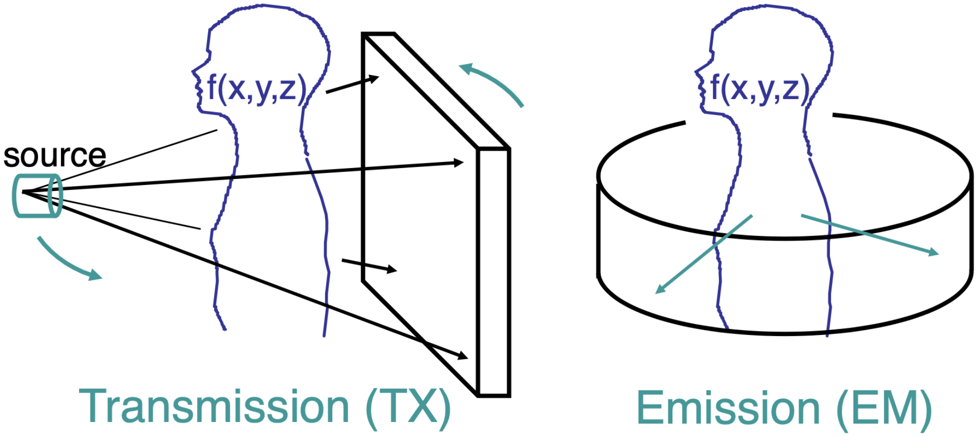
1 detector row: pitch 1 and 2



4 detector rows: pitch 1

Emission Tomography

CT



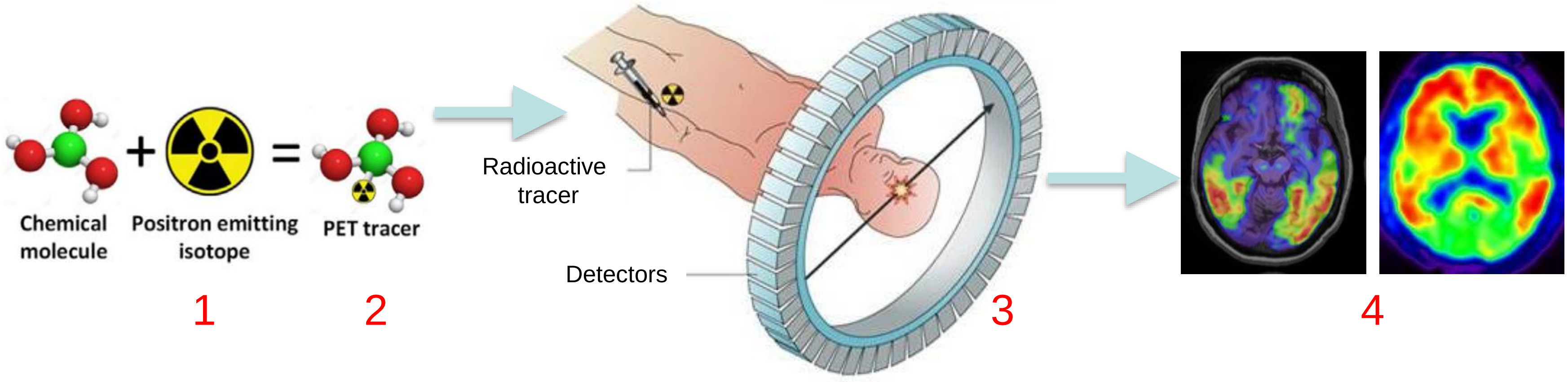
PET

We want photons with differential absorption in tissue and complete absorption in the detectors

We want photons with no absorption in tissue and complete absorption in the detectors

Higher photons energies and thicker detectors for PET

Positron Emission Tomography



1. Radionuclide Production
2. Radiochemistry (Label a tracer)
3. Imaging
4. Data analysis

1. Radionuclide Production

Stable nuclei bombarded with high energy particles

PET RADIONUCLIDES PRODUCED WITH A CYCLOTRON

Nuclide	Half-Life	Nuclear Reaction
O-15	2 min	$^{14}N(d,n)^{15}O; ^{15}N(p,n)^{15}O$
N-13	10 min	$^{12}C(d,n)^{13}N$
C-11	20 min	$^{10}B(d,n)^{11}C$
F-18	110 min	$^{18}O(p,n)^{18}F$

p is proton, n is neutron, d is deuterium (heavy hydrogen with proton and neutron)

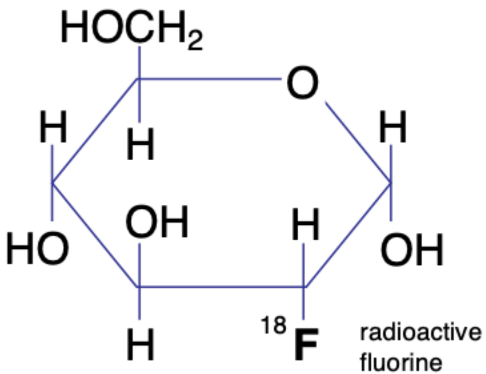
e.g. In production of F-18, O-18 is the target material bombarded with a proton causing a neutron to be emitted in the production of F-18

Requires on-site cyclotron or quick access to isotopes

2. Radiochemistry

Radioactive isotopes are synthesized with biologically meaningful material to form a radiotracer (radiopharmaceutical) - “label” a substance used in the body

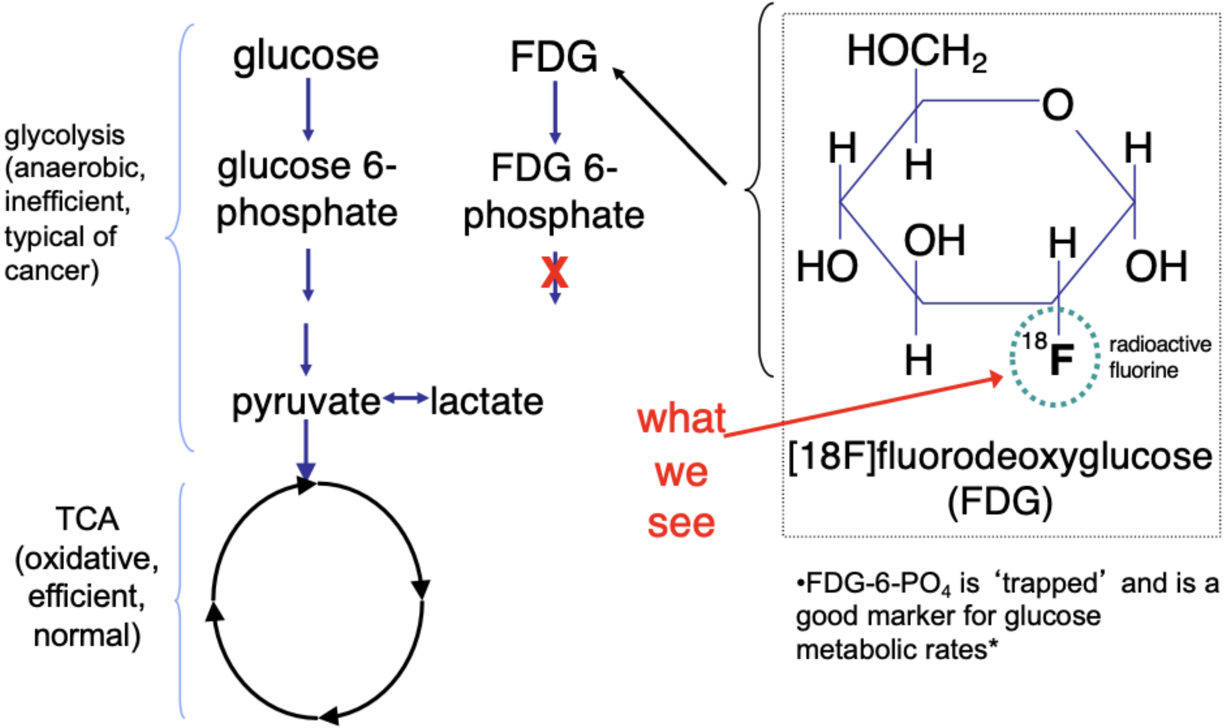
Radiotracer	Injected Dose	Imaging application
[O-15]water	50 mCi	Brain Blood Flow
[N-13]ammonia	25 mCi	Myocardial Perfusion
[C-11]acetate	20 mCi	Myocardial Flow and Metabolism
[F-18]FDG	10 mCi	Brain, Heart, and Tumor Metabolism



[18F]fluorodeoxyglucose (FDG)

3. Imaging

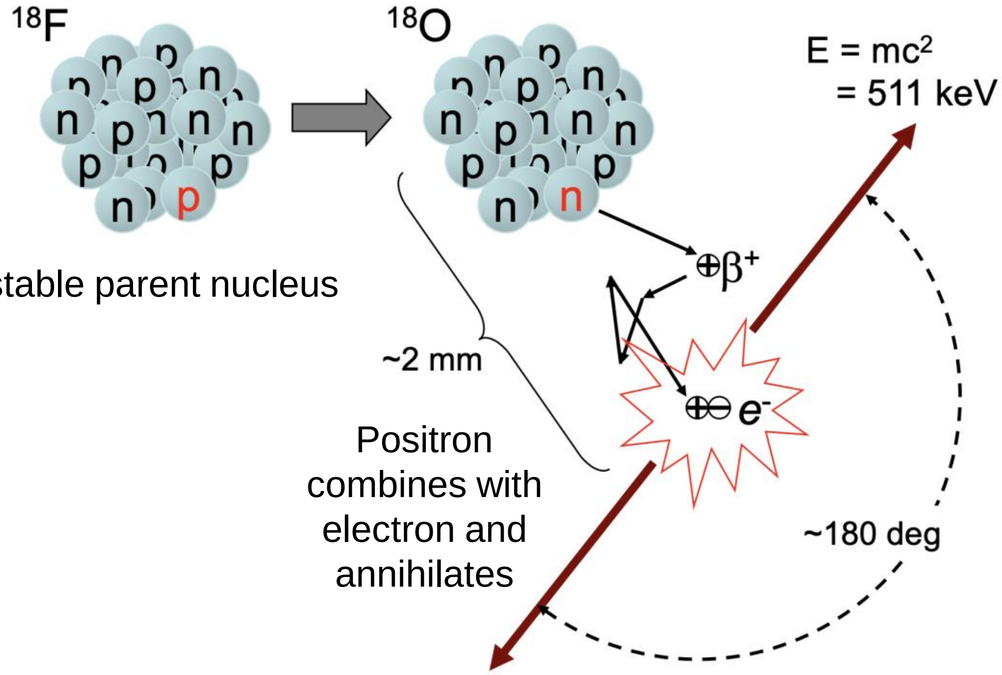
Example: Glucose Metabolism



•$FDG\text{-}6\text{-}PO_4$ is ‘trapped’ and is a good marker for glucose metabolic rates*

3. Imaging

Proton decays to neutron in nucleus



Unstable parent nucleus

Positron combines with electron and annihilates

Two anti-parallel 511 keV photons produced

3. Imaging

Photon travels through subject and interact with matter:

- Photoelectric Absorption
- Compton Scatter

The probability that the photons will continue on a straight line (will not be attenuated) is stated as:

$$P = e^{-\mu x}$$

μ- linear attenuation coefficient, x - distance along a line

Ex. What is the probability a photon will travel on a straight line through 10 cm of adipose tissue?

$$P = e^{-\mu x} = e^{(-0.087/cm)(10cm)} = 0.41$$

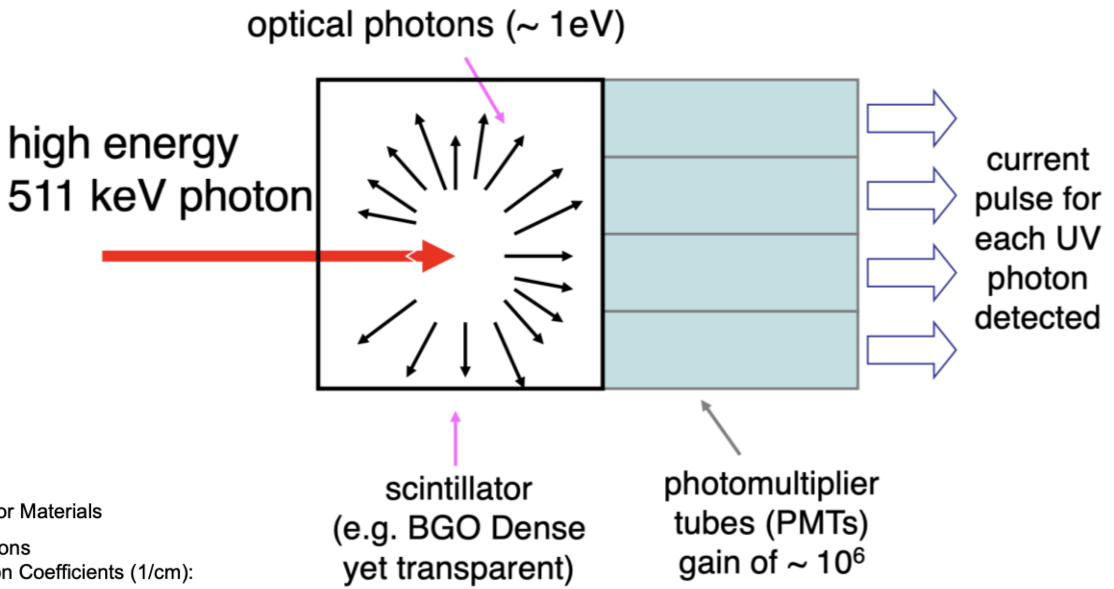
For 511keV photons
Linear Attenuation Coefficients (1/cm):

Body (muscle)	=	0.0932
Adipose (fat)	=	0.0868
Lung	=	0.0267
Wet Spine Bone	=	0.0997
Wet Rib Bone	=	0.1246

3. Imaging

Photons Enter Detectors:

- Photons enter scintillator-(One high energy photon makes many low energy photons)
- Photo multipliers increase light to measurable signal
- Detection system "counts" photon



Ex. What is the probability a photon will travel through 10 cm of BGO?

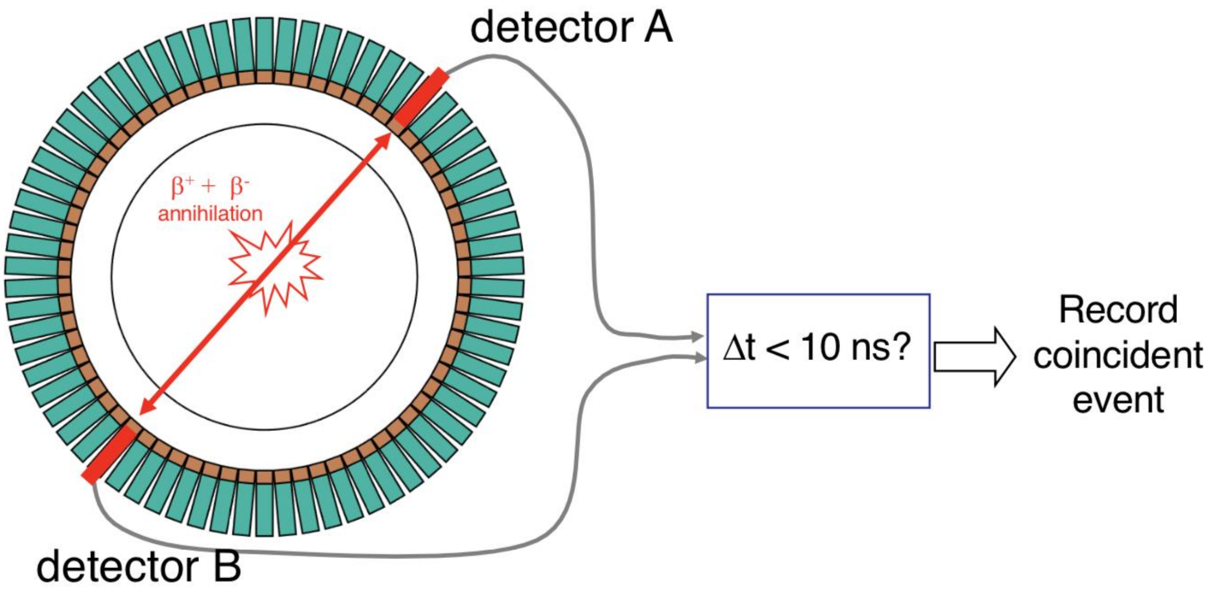
Common Detector Materials

For 511keV photons
Linear Attenuation Coefficients (1/cm):
BGO = 0.955
LSO = 0.833
GSO = 0.674

$$P = e^{-\mu x} = e^{(-0.995/cm)(10\,cm)} = 7.1\times10^{-5}$$

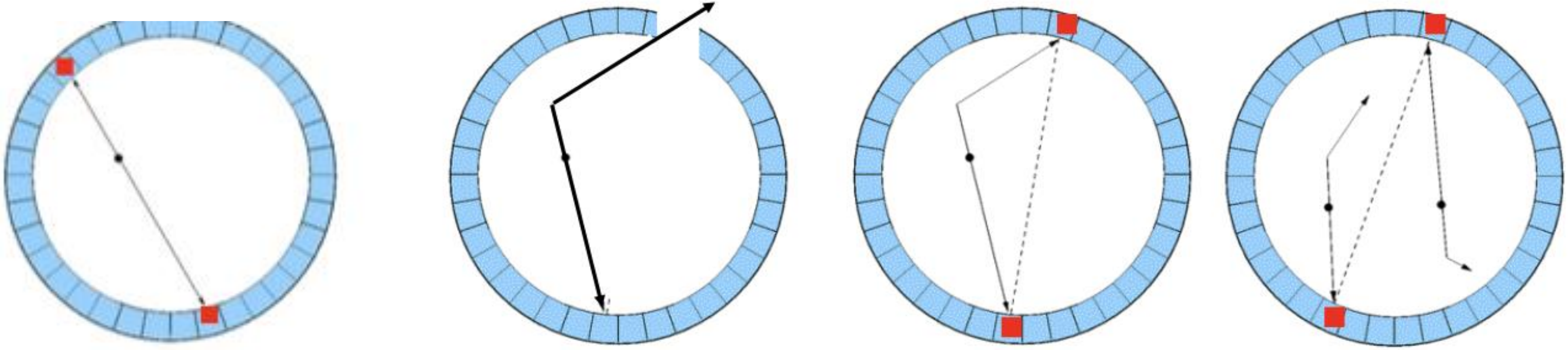
3. Imaging

Photons pairs into coincident events



3. Imaging – Challenges

Errors in line of response (LOR) measurements

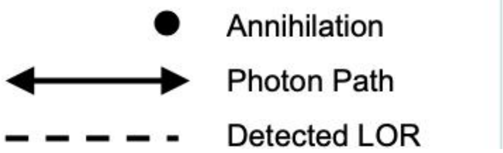


True Event
Both annihilation photons escape without scatter and interact in detector

Lost (Attenuated) Event

Scattered Coincident Event

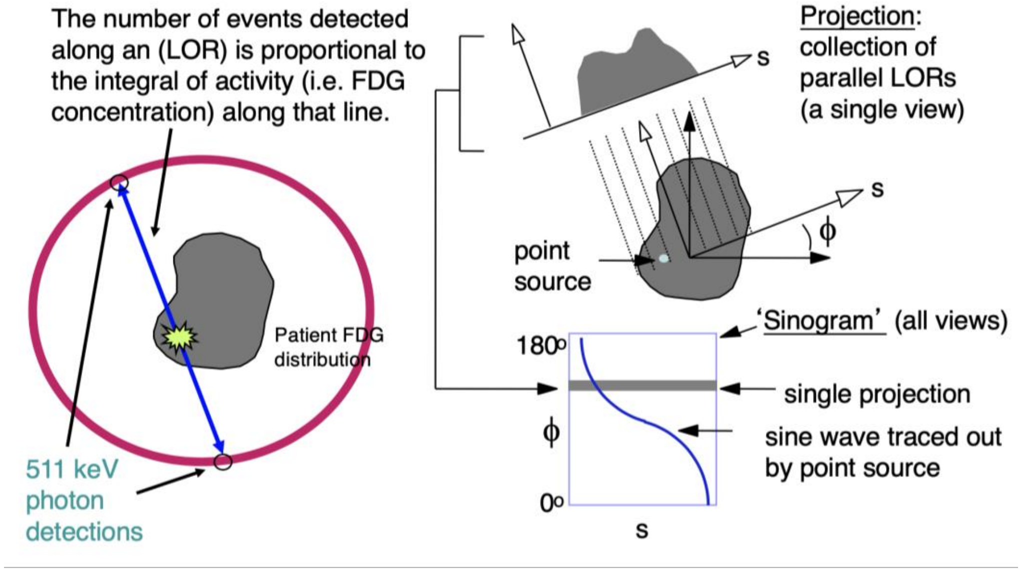
Accidental Event (Random)

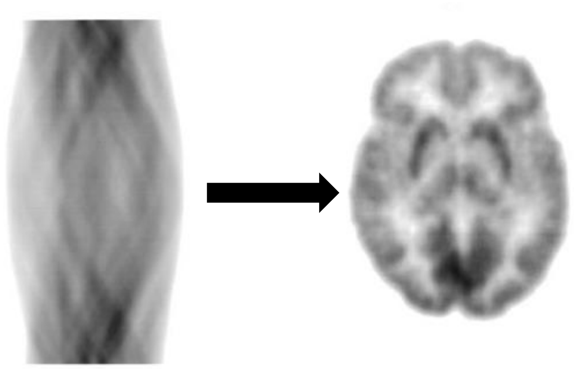


4. Data analysis

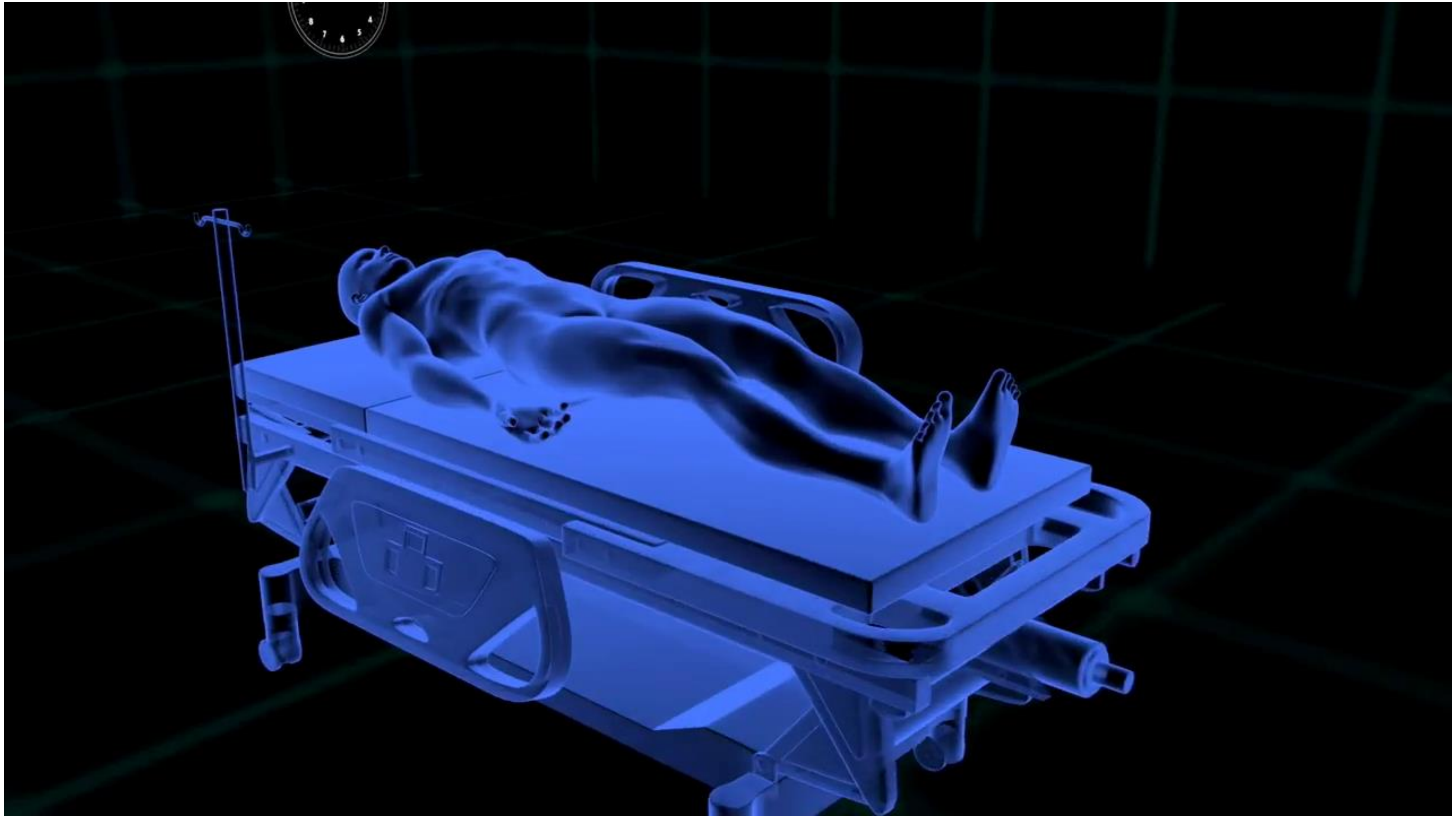
Data organized into sinograms
LORs => Sinogram

Sinograms reconstructed into image
(Methods: FBP, Iterative,…)

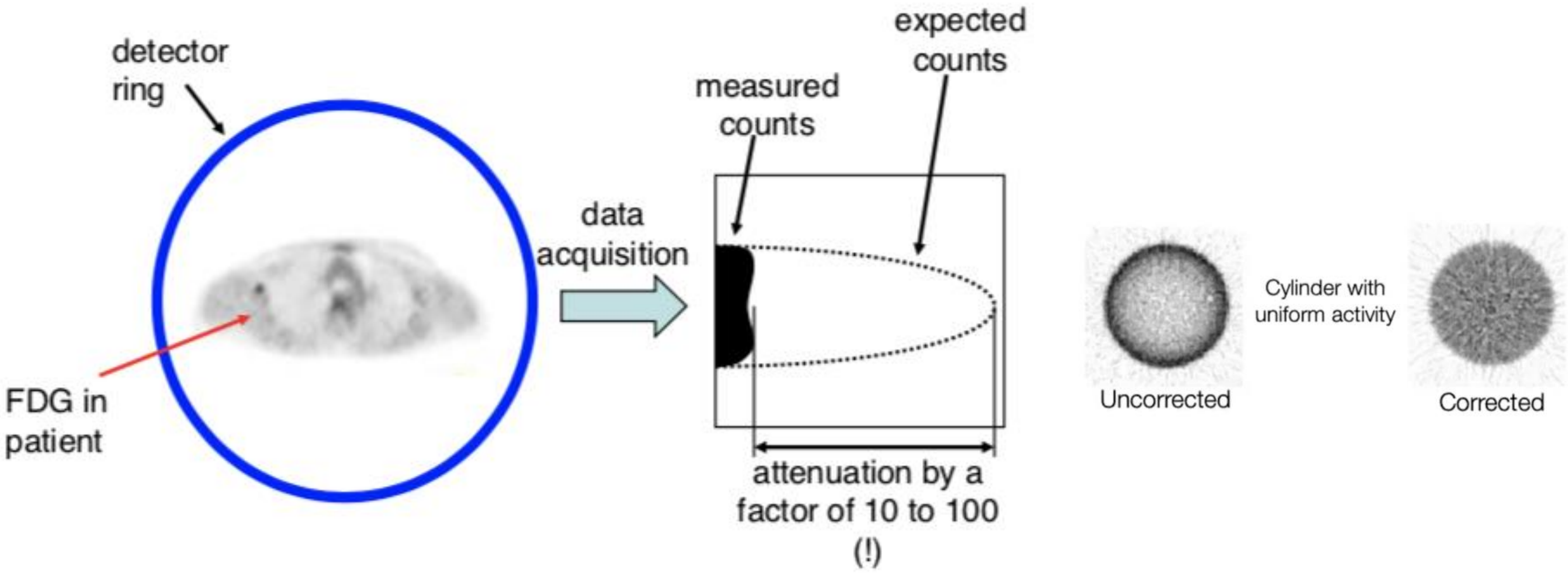




Positron Emission Tomography - Summary

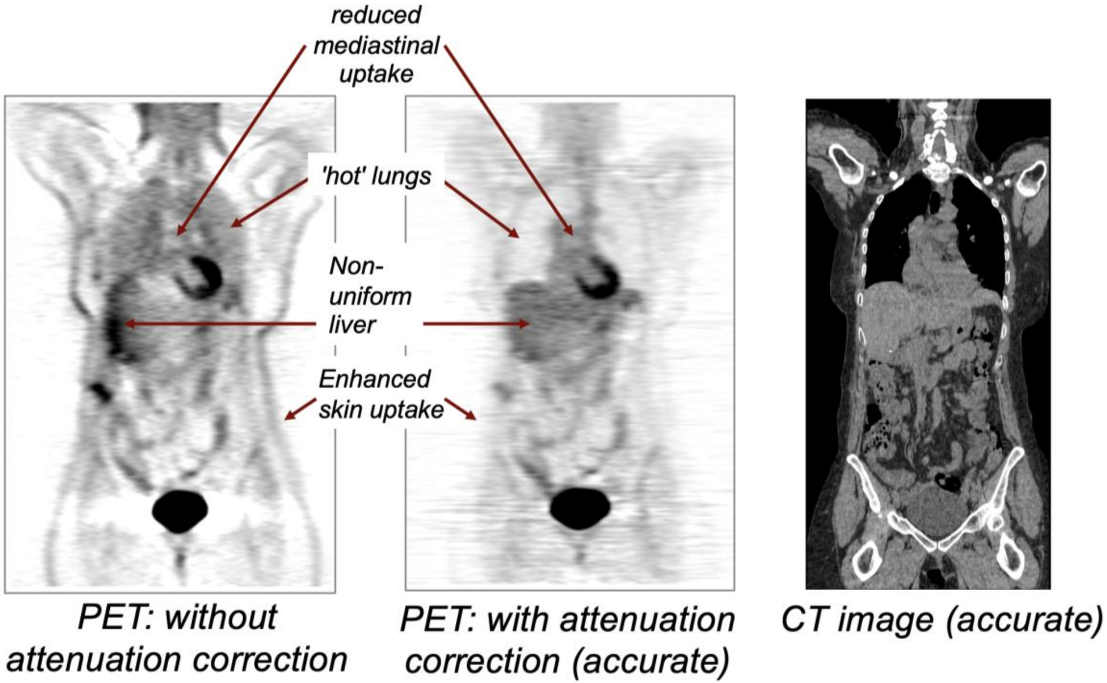


4. Data analysis – correcting for errors



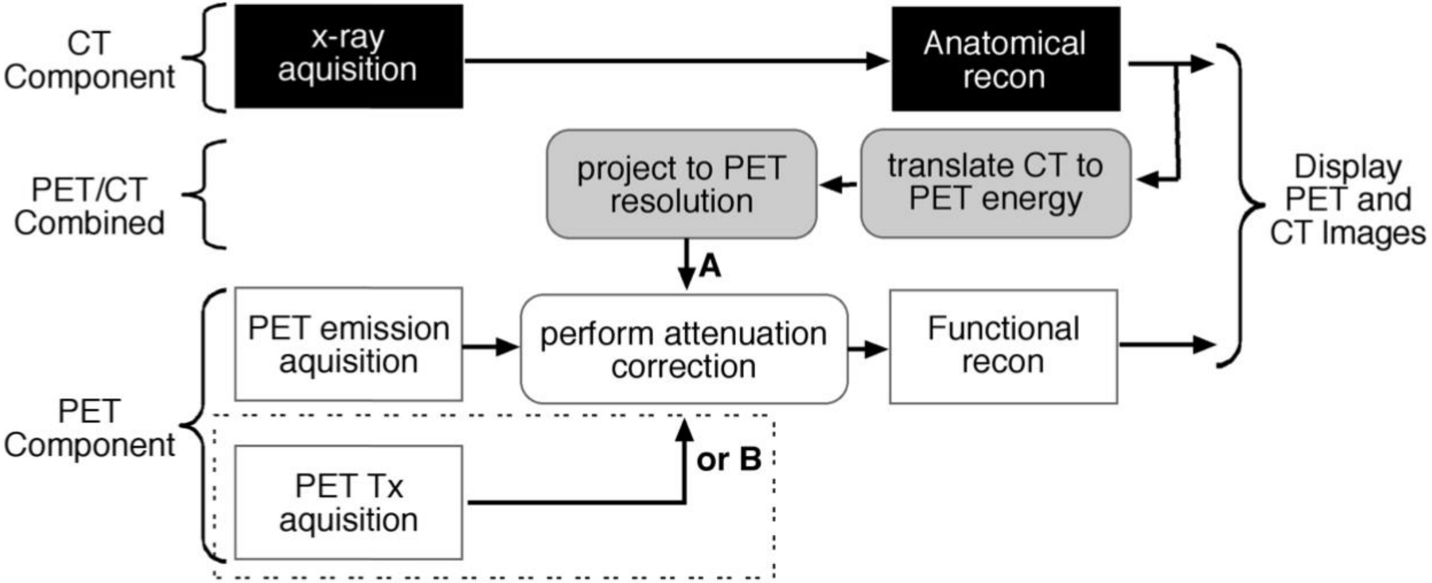
- Attenuation is mainly due to Compton scatter
- It is by far the most important effect for both noise (due to reduced counts) and qualitative image appearance

Effects of attenuation



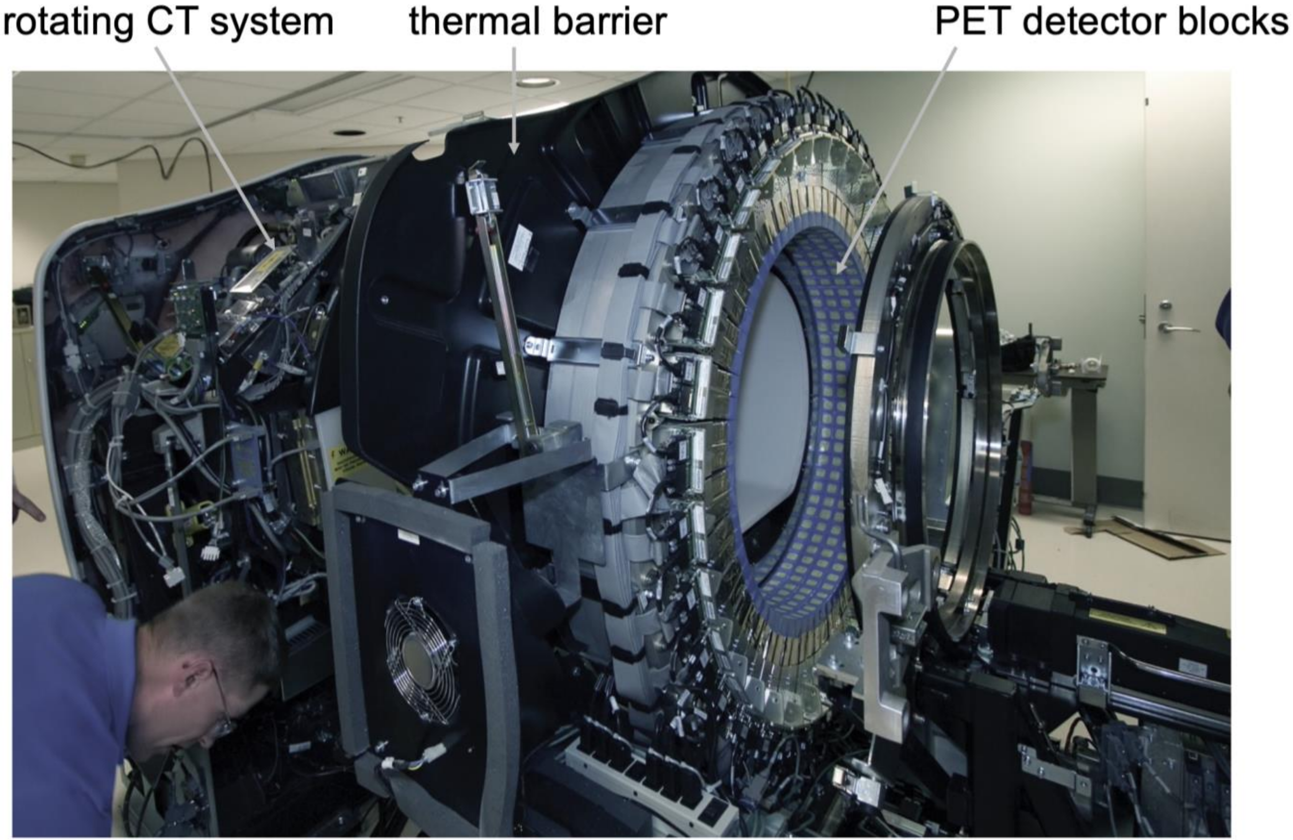
- Attenuation can be estimated given:
  - ✓ Total distance traveled in object
  - ✓ Attenuation coefficient
- Can be performed with a CT image

Flowchart of typical PET/CT operation

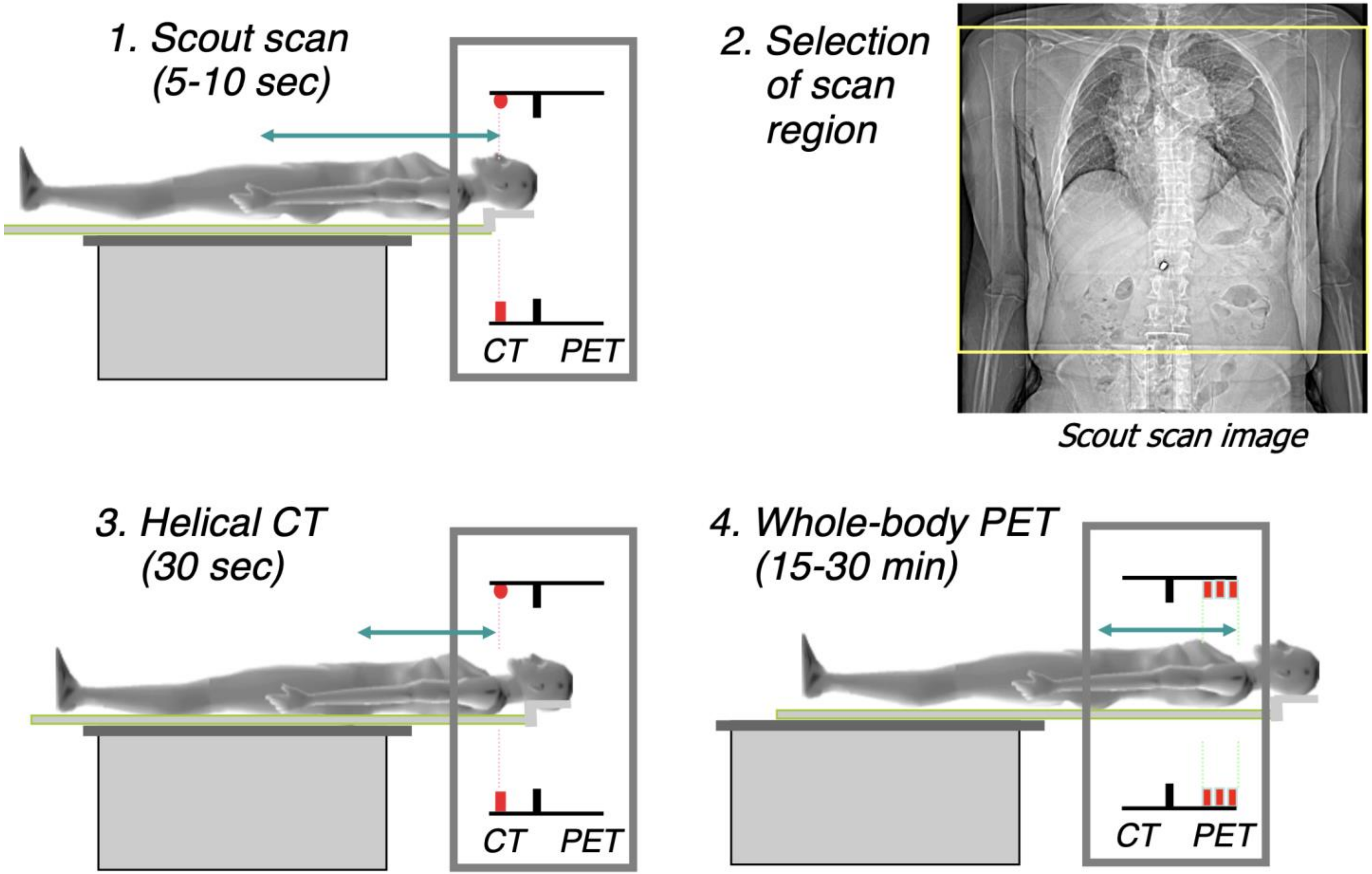


Attenuation correction factors can be obtained from a CT image

Commercial/Clinical PET/CT Scanner

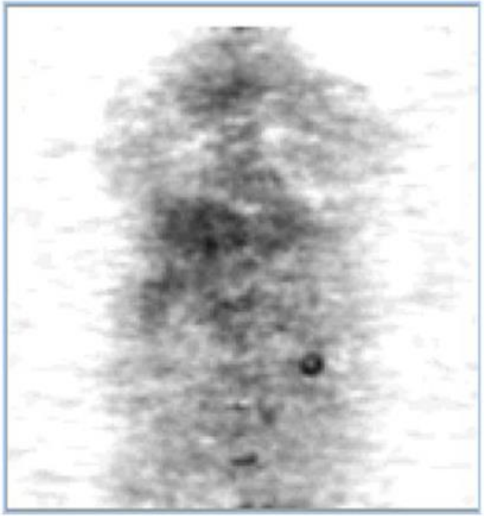


Typical PET/CT Scan Protocol



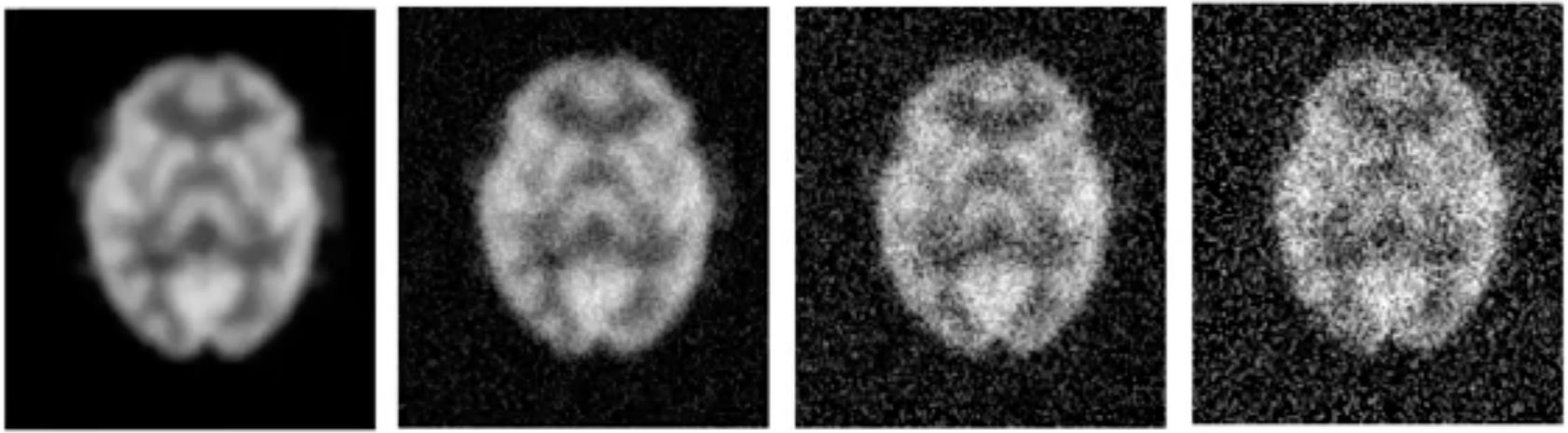
Imaging physiology (FDG-PET) & Anatomy (CT)

Image Quality Assessment



Question: Which one is a better image?

Trade-offs in selecting imaging parameters



High Sensitivity	Low Sensitivity
Long Scan	Short Scan
High dose	Low dose

Trade-offs in selecting imaging parameters

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

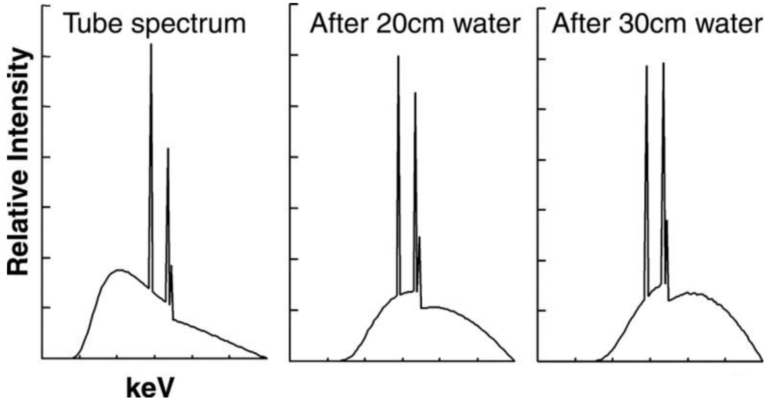
- kVp not only controls the dose but also controls other factors such as image contrast, noise and x-ray beam penetration through patient

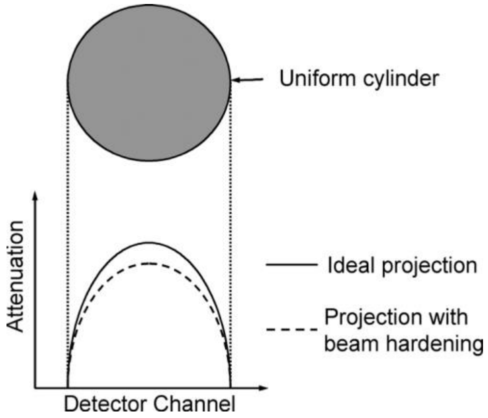
Questions?

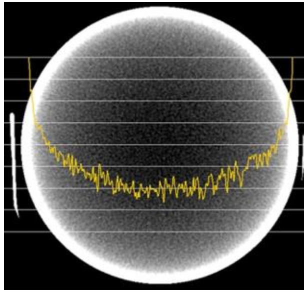
Special Thanks to Paul Kinahan and Adam Alessio for sharing their slides

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

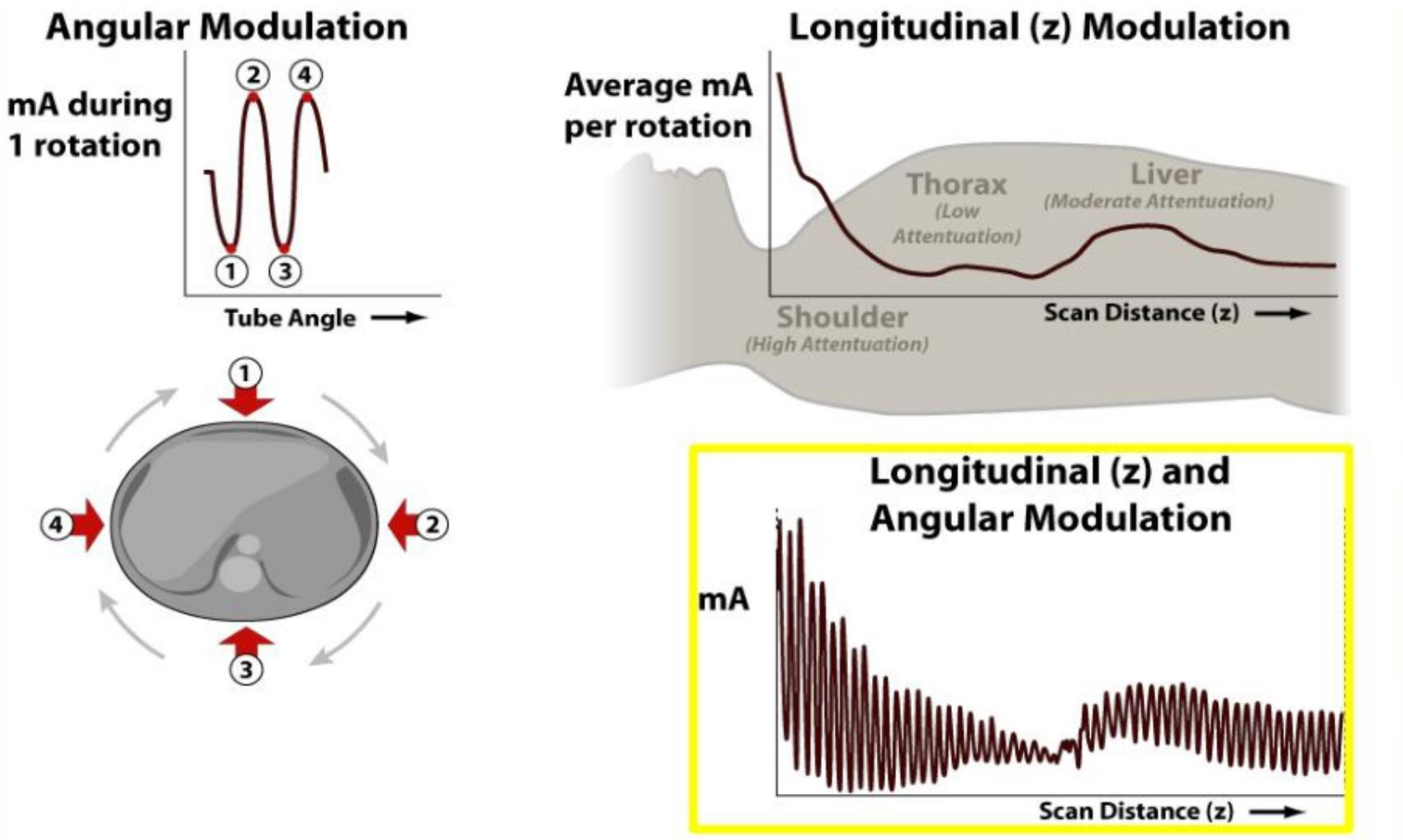
Beam Hardening Correction

- Filtration: A flat piece of attenuating, usually metallic material is used to “pre-harden” the beam by filtering out the lower-energy components before it passes through the patient. An additional “bowtie” filter further hardens the edges of the beam, which will pass through the thinner parts of the patient.

- Calibration correction: Manufacturers calibrate their scanners using phantoms in a range of sizes. This allows the detectors to be calibrated with compensation tailored for the beam hardening effects of different parts of the patient.

- Beam hardening correction software

Attenuation versus body location and direction



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