Physical Principles of SPECT

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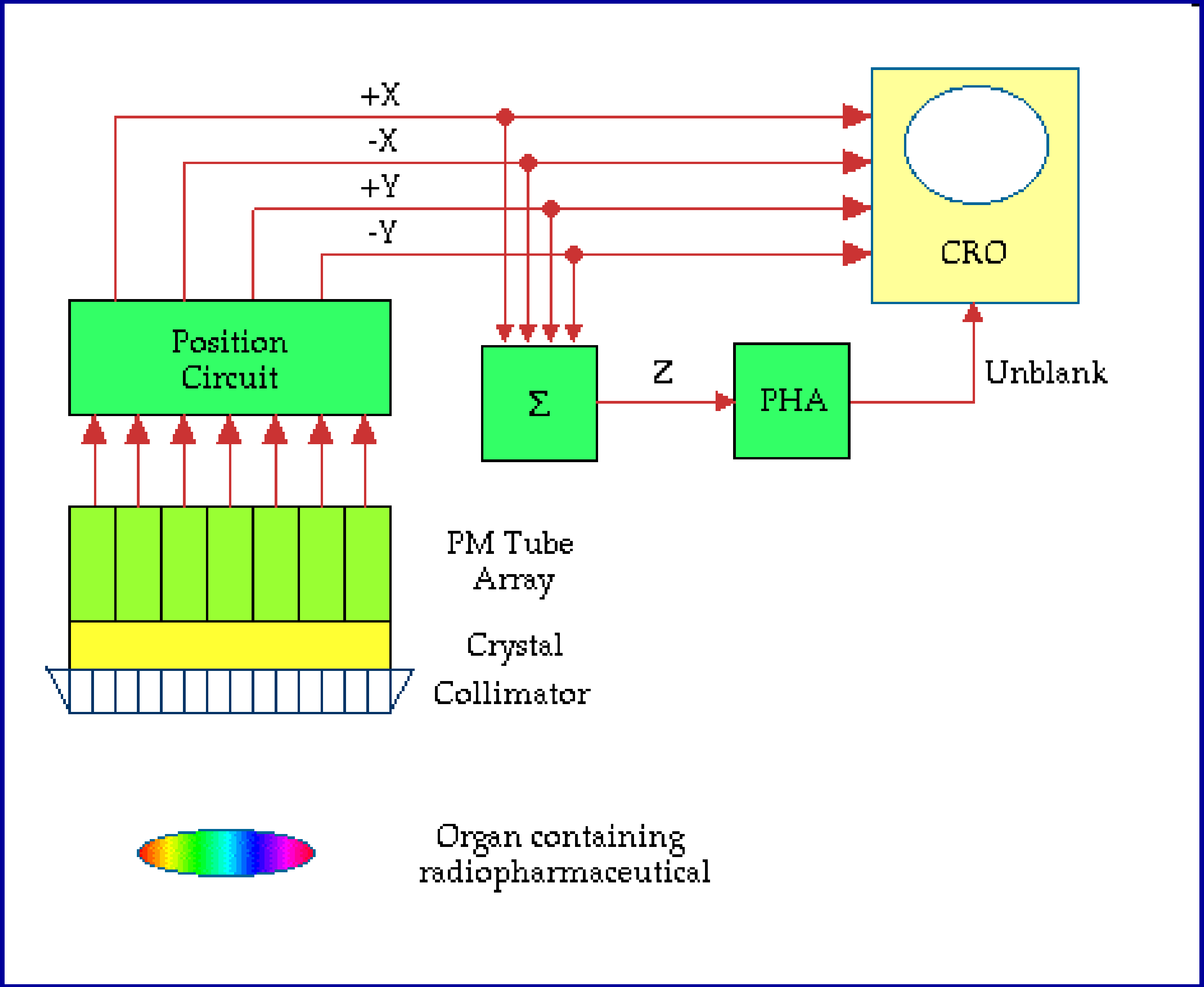
SPECT Basics

Single Photon Emission Computed Tomography

- Radiopharmaceutical administration (injected, ingested, or inhaled)
- Biodistribution of radiopharmaceutical
- Gamma camera detects gamma rays and images (tomography) the distribution of radiopharmaceutical within the patient
- Used for visualization of functional information

Anatomy of gamma camera



SPECT Imaging

- Advantages of SPECT
  - Improved image contrast
  - Improved quantification of cardiac function, tumor/organ volume determination, the quantification of radioisotope uptake
  - Problems of gamma-ray attenuation and scatter may be better handled by SPECT (although, as yet, not completely), over planar projection imaging

SPECT acquires 2-D projection of a 3-D volume

Improvements in SPECT technology

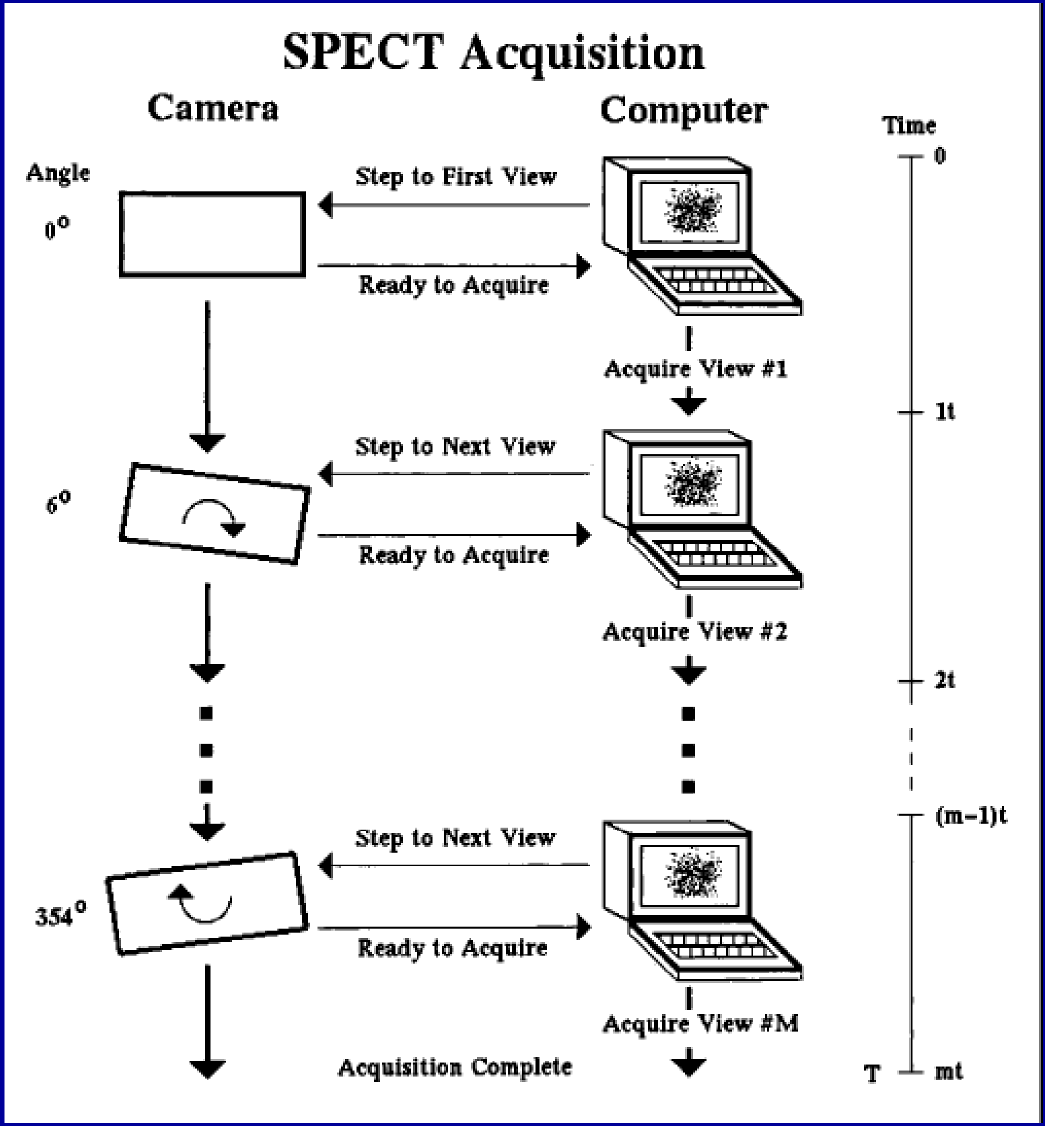
- The application of multiple gamma camera heads
- Noncircular orbits
- The application of non-uniform attenuation correction methods
- Gated SPECT perfusion scans with 99mTc agents and 201-Tl, also gated SPECT blood pool

SPECT data acquisition

- *Arc of Rotation*
- *Matrix size*
- *Radius of Rotation and Number of Projections*

Arc of Rotation

- Step-and-shoot mode
- Continuous mode



Why opposing views are not the same in gamma camera imaging

- Resolution degrades as distance from camera increases
- Certain percentage of Compton scatter is accepted as photopeak gamma rays, due to the finite energy resolution of the camera
- Photon attenuation

Therefore, 360° of arc is required for accurate reconstruction in most SPECT studies

Matrix size

- The 2 matrix sizes associated with SPECT imaging are 64 x 64 and 128 x 128

6.25 mm 3.125 mm

Size of a pixel should, ideally, be less than 1/3 of the expected (FWHM) resolution of the SPECT system, measured at center of rotation for the isotope being imaged

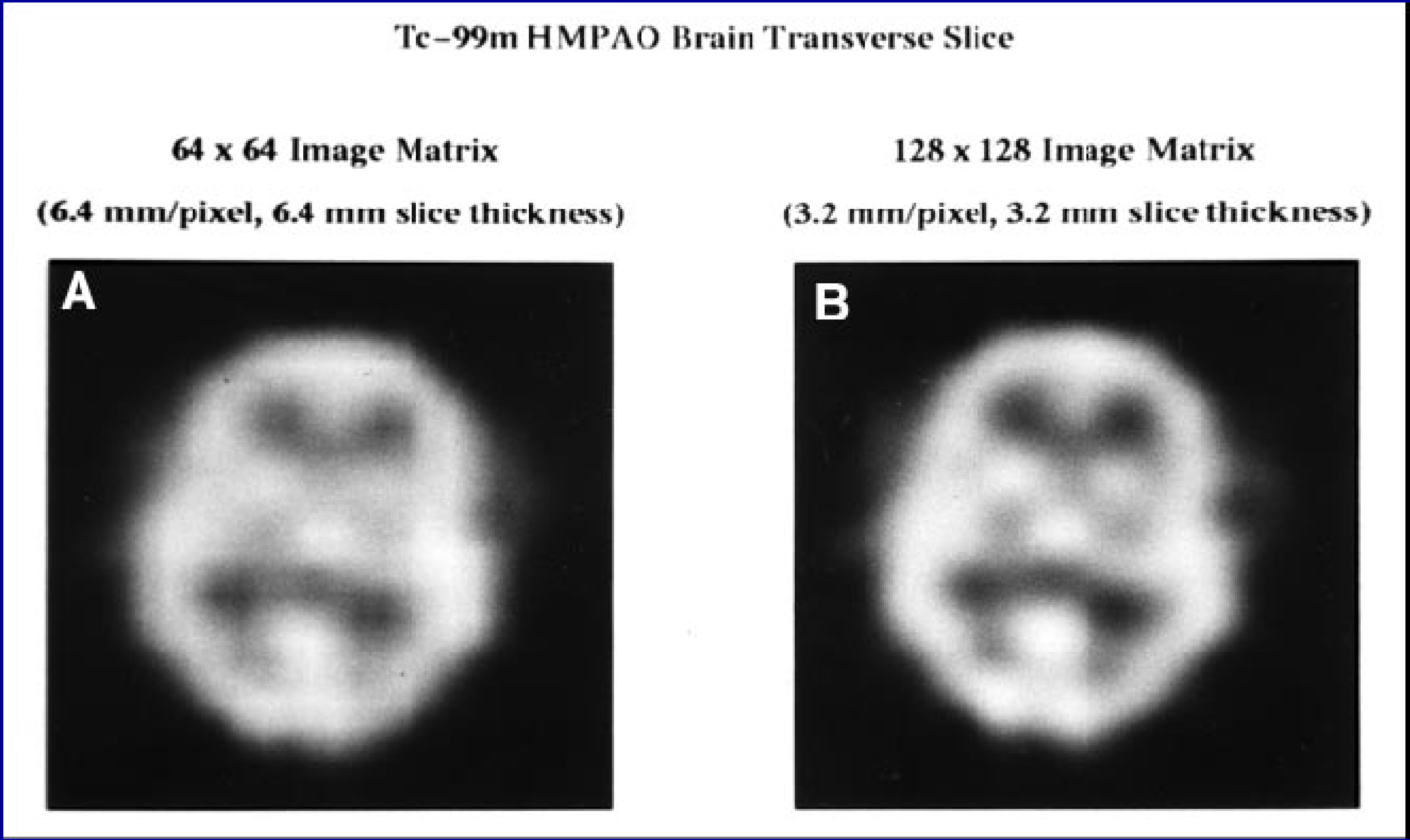
$D = FOV/(Z \times n)$

FOV (mm) = the widest dimension of the computer image matrix

Z = zoom factor (e.g., 1.5, 2.0, etc.) during acquisition

n = number of pixels (e.g., 64 or 128)

Matrix size


Tc–99m HMPAO Brain Transverse Slice
64 x 64 Image Matrix
(6.4 mm/pixel, 6.4 mm slice thickness)
128 x 128 Image Matrix
(3.2 mm/pixel, 3.2 mm slice thickness)
A
B

Matrix size vs Noise

% rms noise = $(120 \times (v)^{3/4})/(N)^{1/2}$

V= the total number of voxels covering the reconstructed object, based on a circular FOV

N= total number of events acquired

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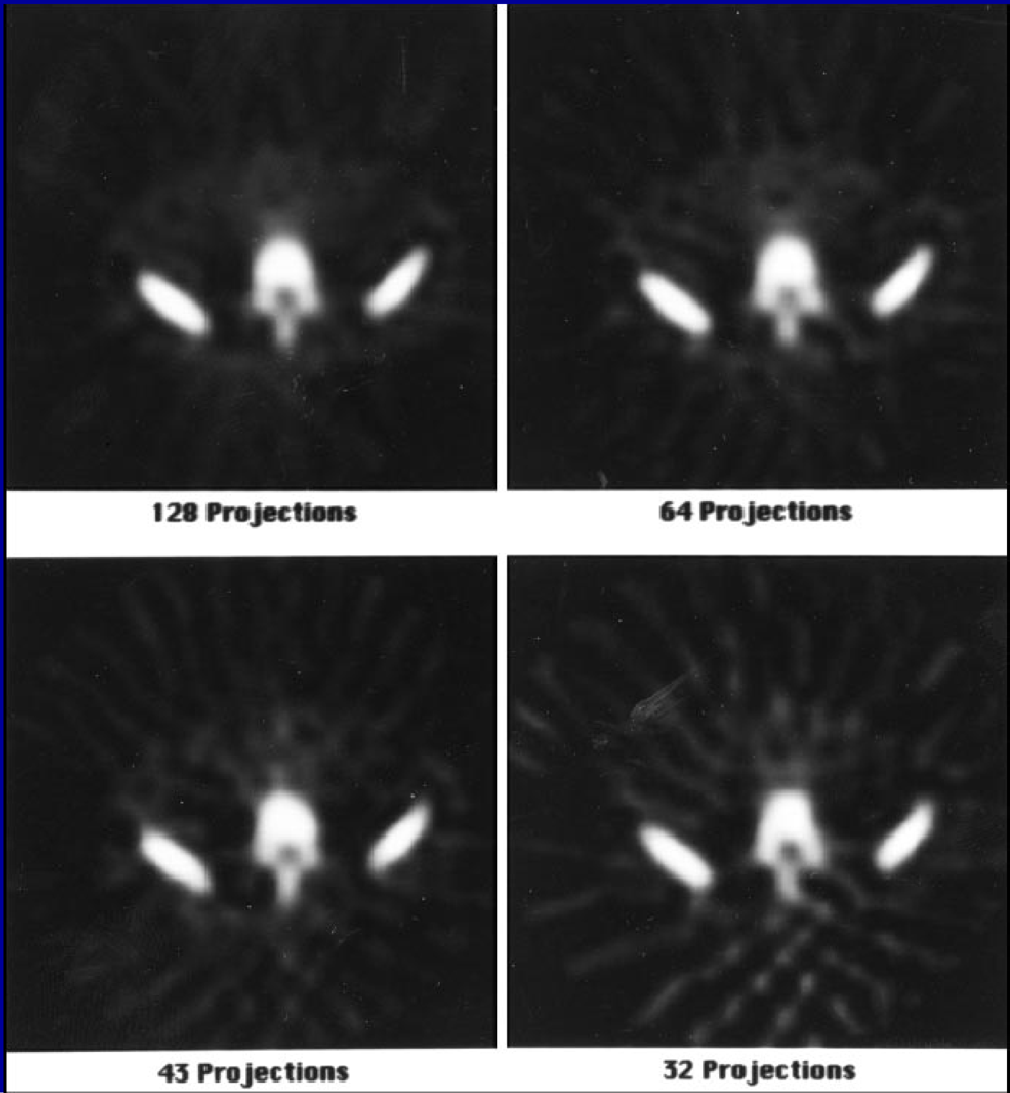
The 128x128 reconstructed percent noise of 128 will ~ be 3 times that of the 64 reconstruction (need to increase count 4 times in the projection data)

Radius of rotation and number of projections

For accurate reconstruction, the number of angular views for 360° should be at least equal to the projection image matrix size

Streak artifacts →



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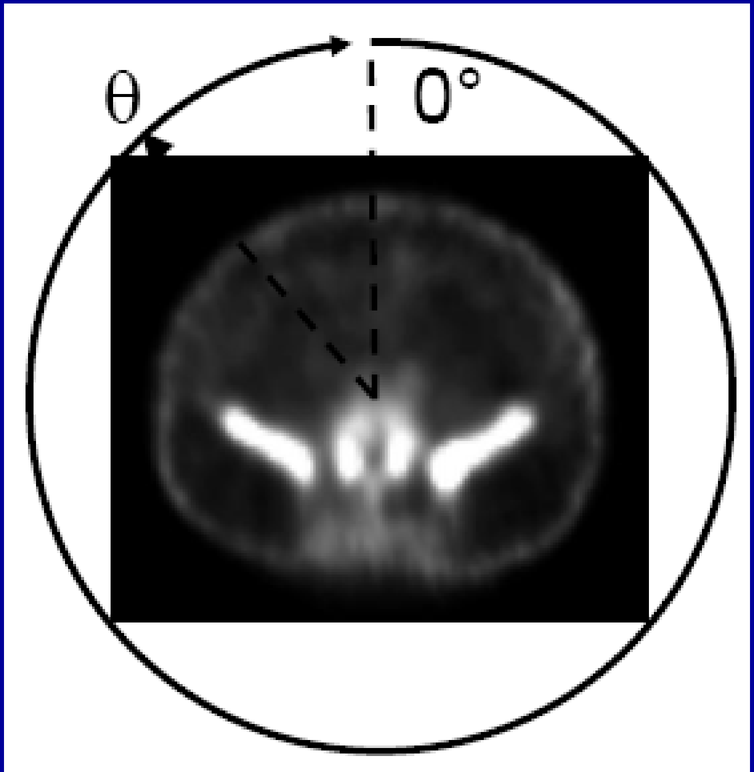
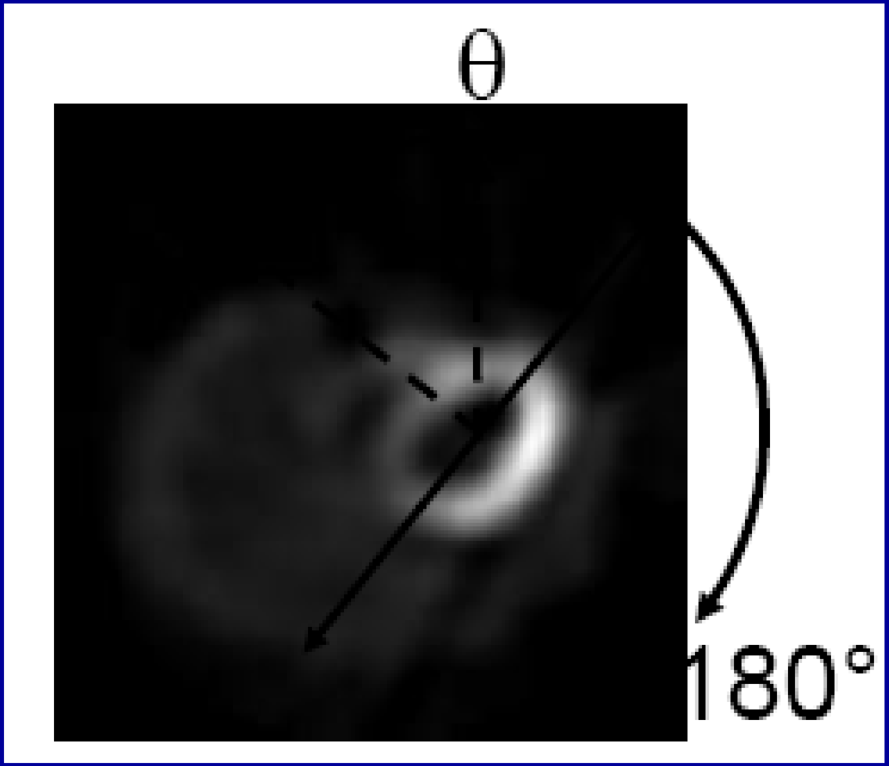
Projections acquired over 360° vs. 180°



Bone imaging



cardiac imaging

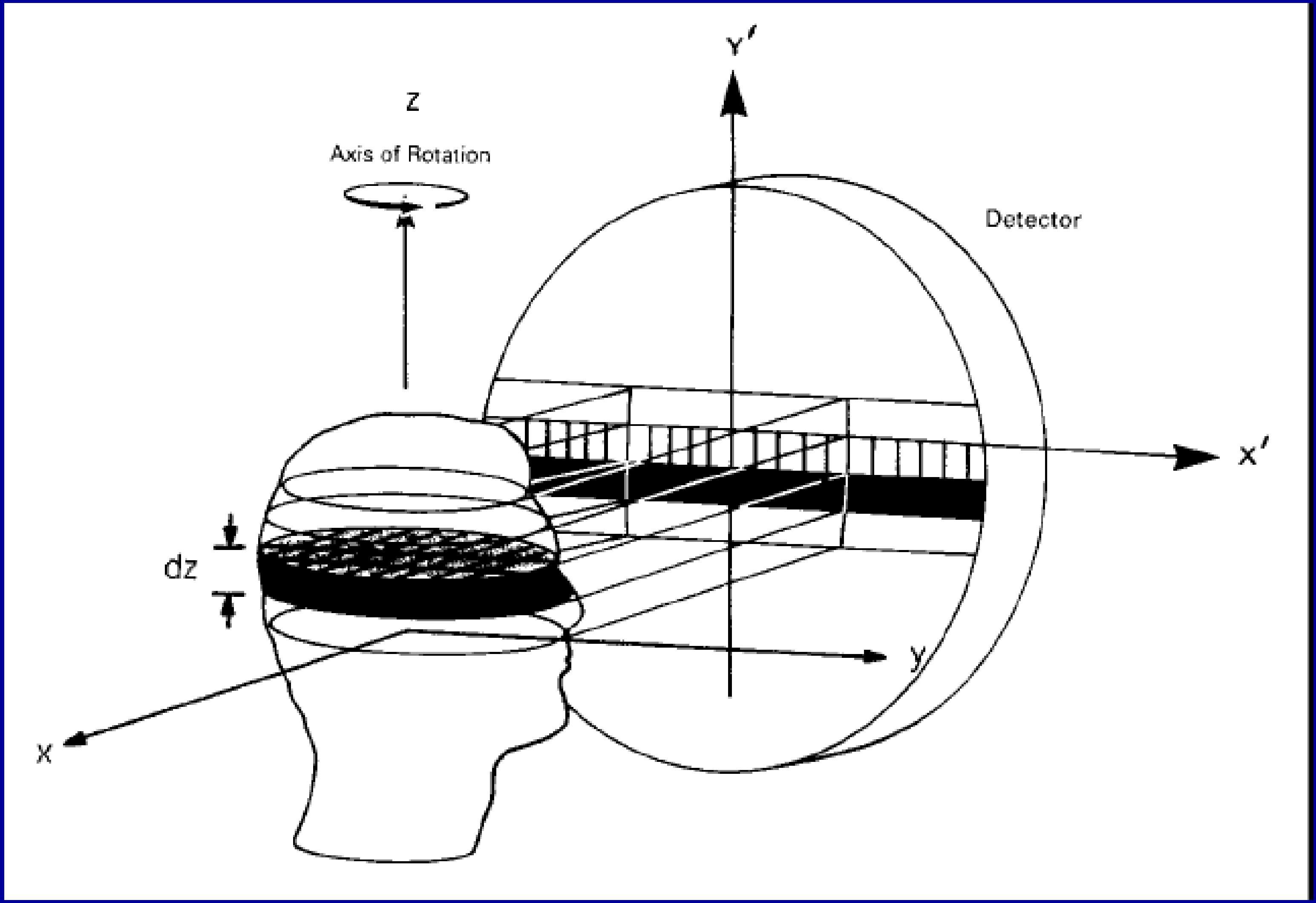
Non-circular vs Circular orbit

- In NCO, the camera will move in and out radially as it rotates around the patient
- NCO results in varying resolution between projections
- May create an artifact (lateral wall of the heart is more intense than septum)

Principles of image reconstruction

- *Filtered backprojection technique*
  - *Radon Transform*
- *Iterative reconstruction techniques*
  - an estimate of the actual *probability* of a certain amount of radioactivity at a particular location being detected by the imaging system at each particular point in each projection.

SPECT projection imaging process (radon transform)



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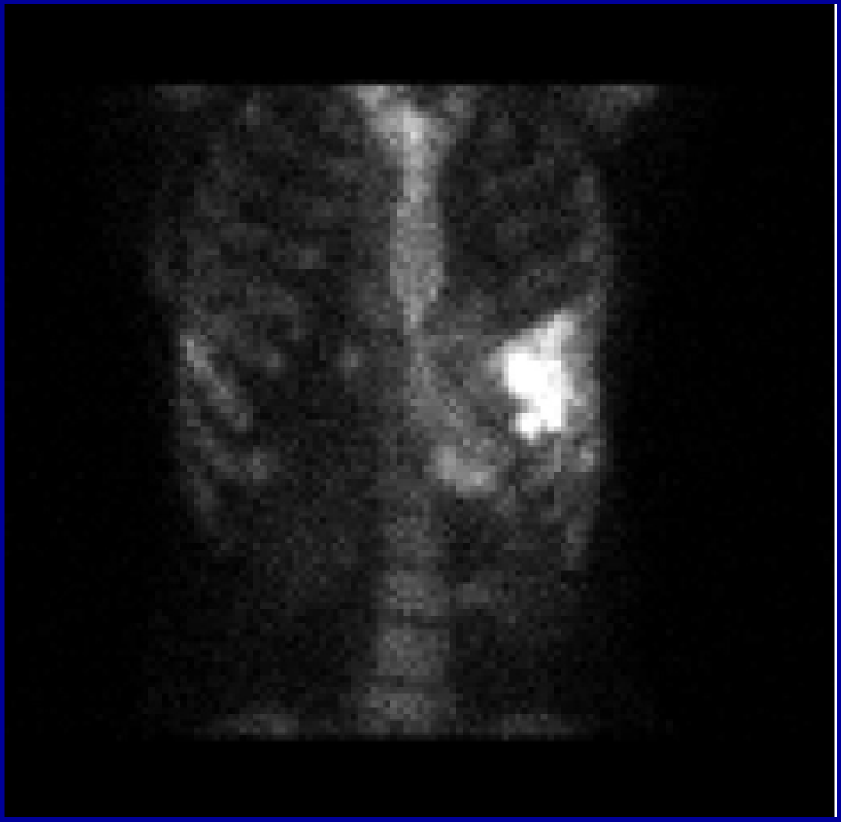
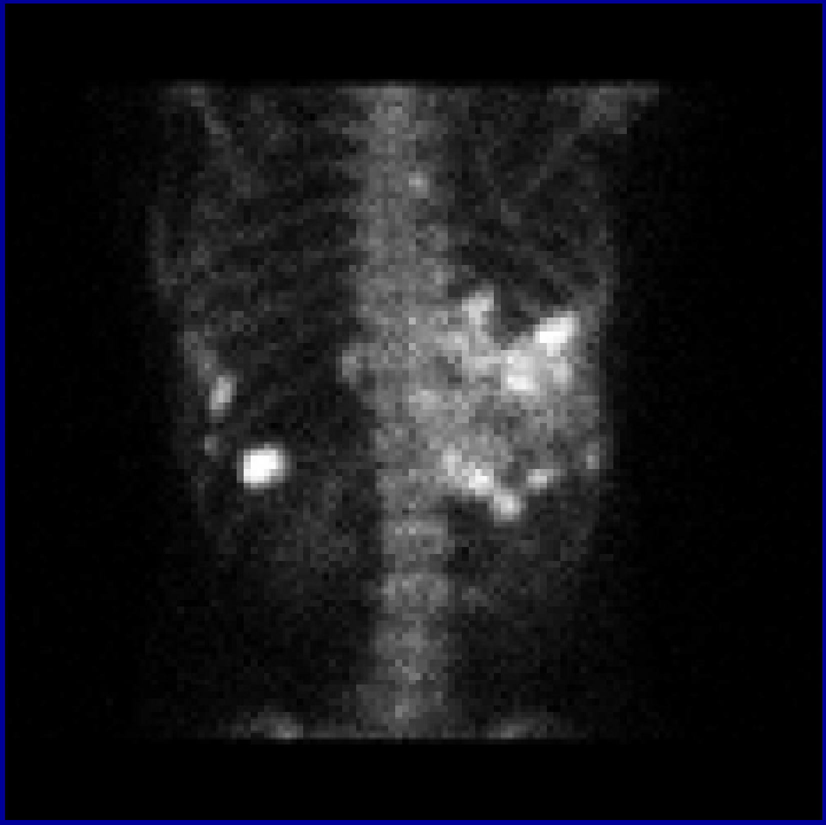
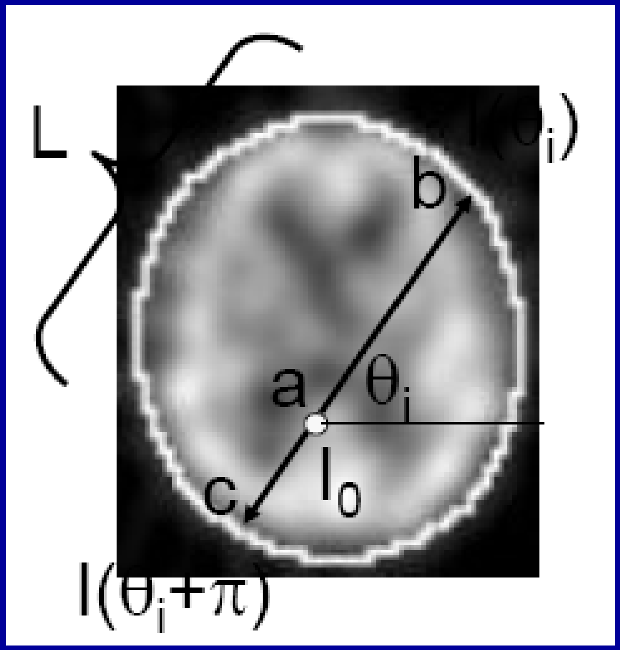
Radon transform angular symmetry is not achieved in SPECT

$P(\theta) \neq P(\theta + \pi)$





FBP assumes a linear & angular symmetry of projections



Differential attenuation

Depth dependant

$$I(\theta_i) = I_0 \, e^{-\int_a^b \mu(L)dL}$$

$$I(\theta_i+\pi) = I_0 \, e^{-\int_a^c \mu(L)dL}$$

Fourier Transform

- Spatial information is converted to *frequency* by *Fourier transform*
- What spatial frequencies are contained in SPECT images?
  – ability to detect high frequencies (small objects)
  – how finely the data is sampled (no of pixels)
    - Sampling Theorem

F_N is the Nyquist frequency

D is the pixel size

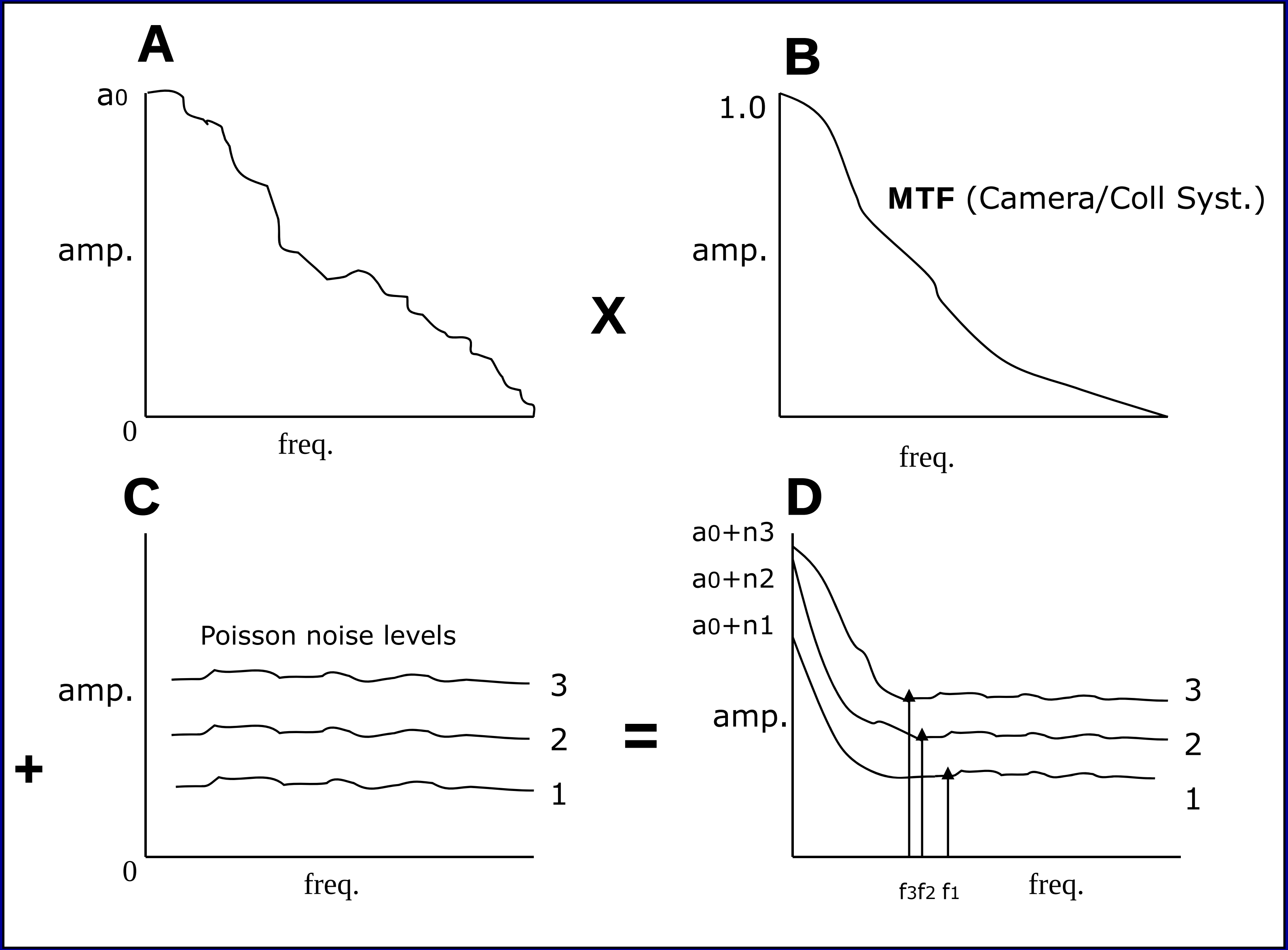
$$F_N = \frac{1}{(2 \times D)}$$

F_N = 0.16 /mm for 128x128

F_N = 0.08/mm for 64x64

Filtered backprojection

- The profiles must first be filtered by a linear ramp in the spatial frequency domain
- The linear ramp increases amplification as the frequencies increase to remove the blurring effect of the projection process
- only a smoothed version of the original 3-D distribution at best may be reconstructed to begin with


A
a0
amp.
0
freq.
X
B
1.0
MTF (Camera/Coll Syst.)
amp.
freq.
C
+
Poisson noise levels
amp.
3
2
1
0
freq.
=
D
a0+n3
a0+n2
a0+n1
amp.
3
2
1
f3f2 f1
freq.

Reconstruction filters

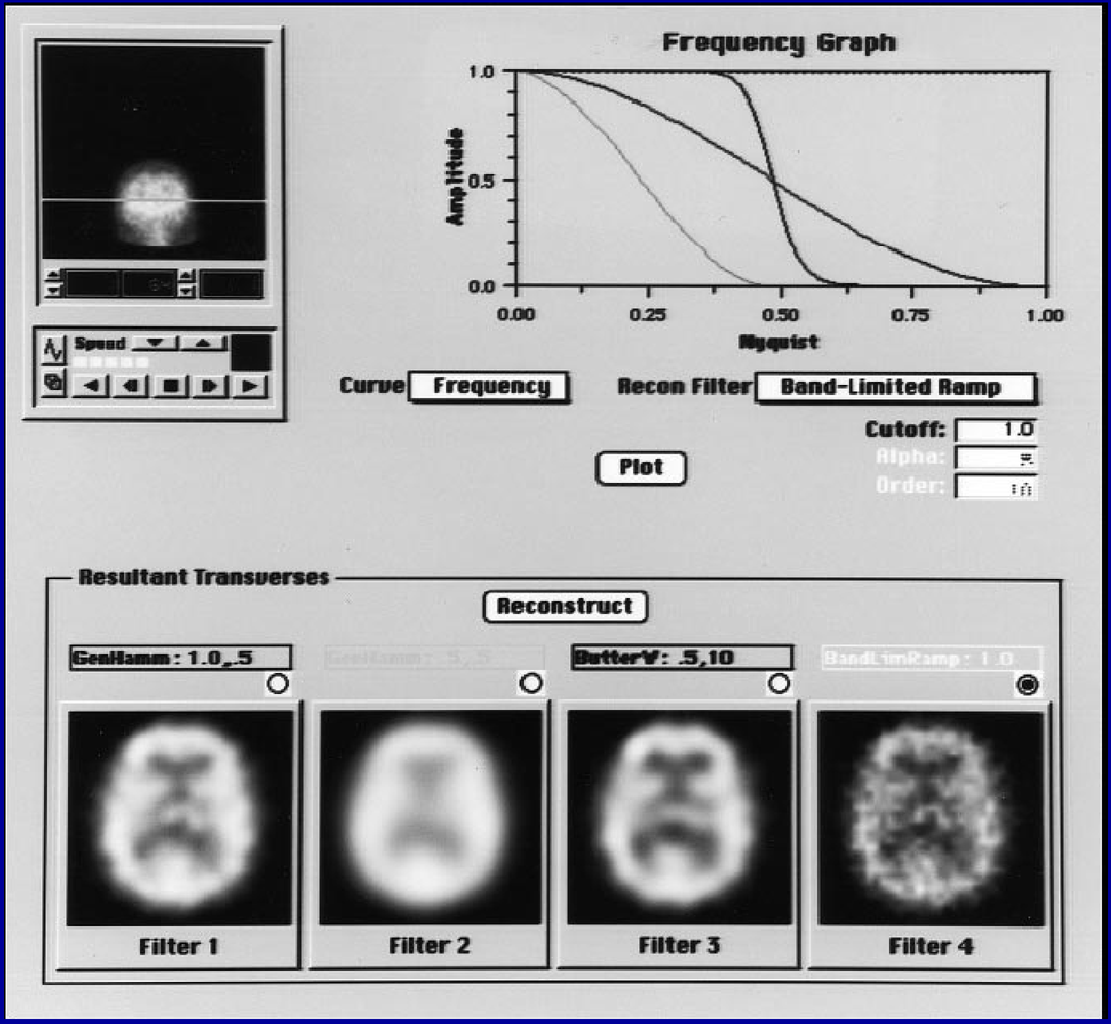
- Cut off linear ramp around the point where patient data frequencies disappear into noise

Standard SPECT Reconstruction Window Filter Functions

Filter	Filter variables
Band-limited ramp	Cut-off frequency
Shepp-Logan	Cut-off frequency
Shepp-Logan-Hanning	Cut-off frequency
Generalized Hamming	Cut-off frequency, a weighting coefficient
Low-pass cosine	Cut-off frequency
Butterworth	Cut-off frequency, order
Parzen	Cut-off frequency

The effect of reconstruction filter on image noise/resolution

In Butterworth filter, as the order increases, the function has steeper roll off around the cut off frequency, allowing better adaptation of the filter function to frequency characteristics



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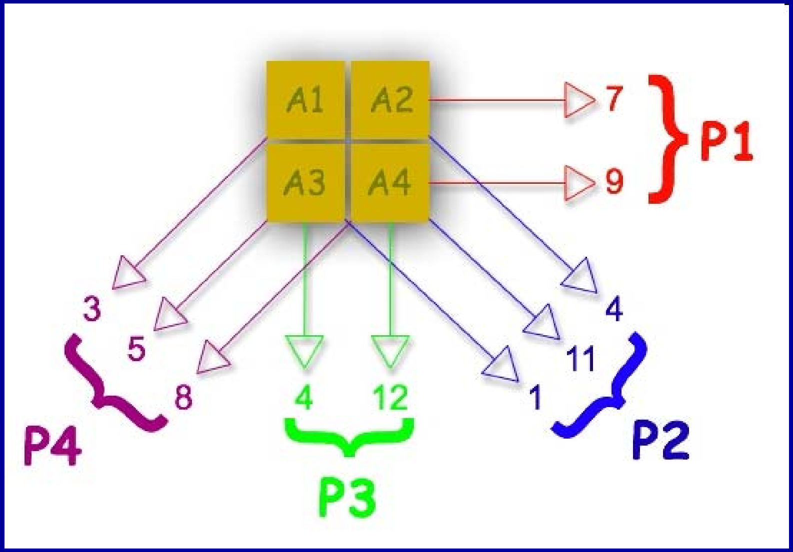
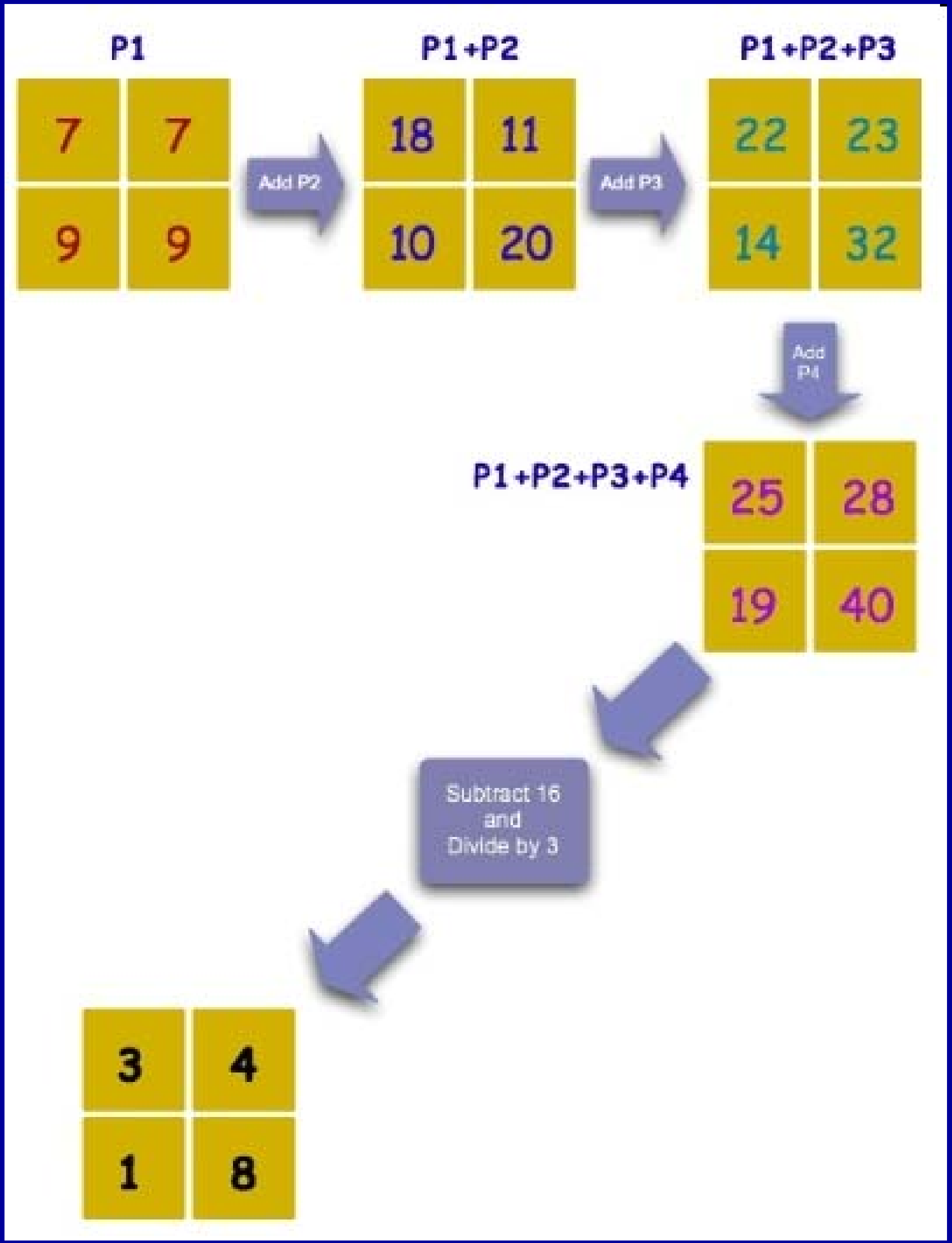




- Backprojection

Iterative Reconstruction Methods

- Became a requirement for accurate attenuation correction based on transmission sources scan data
- CPU with high speed required to perform intense computations for reconstruction
- Incorporates physics of gamma camera in reconstruction model

Iterative Reconstruction Methods

- Step1:
- Calculating a probability density matrix (PDM_i) for each voxel $F_{i..}$
- Each data point D_{ij} in PMD_i, represents the probability that radioactivity at voxel F_i will be detected in pixel P_j in projection set.
- Each D_{ij} in each PDM_i is calculated using attenuation map & resolution functions

Iterative Reconstruction Methods

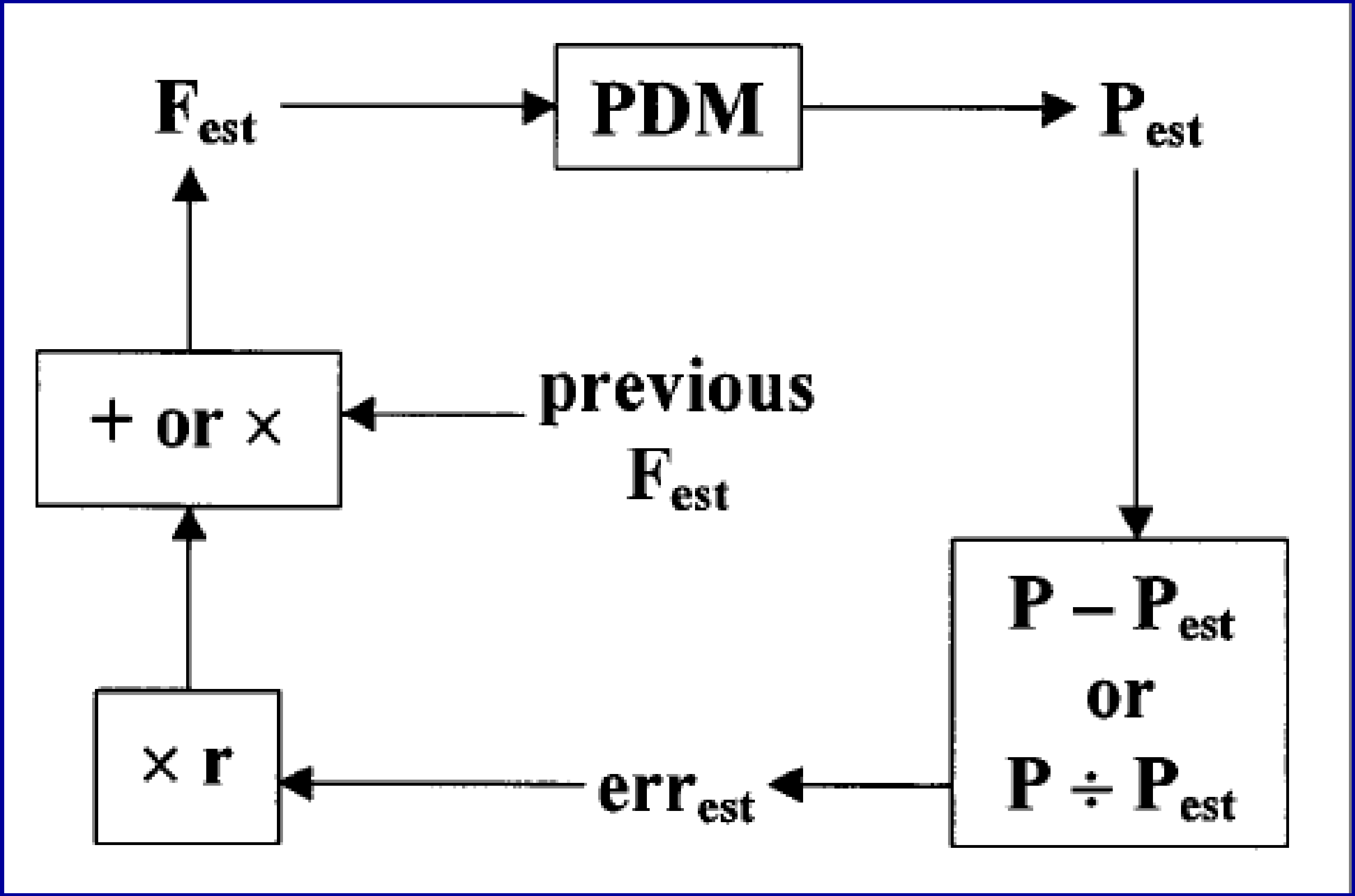
- Step2:

➢ Calculation of an initial estimate of reconstructed slice (same as fbp)

➢ Repeated or iterative, execution of projection/backprojection process using PDMs

➢ Until a desired number of iterations has been reached

Principle of iterative reconstruction



Iterative methods: the MLEM algorithm (maximum-likelihood expectation maximization) and ordered-subset EM (OSEM),

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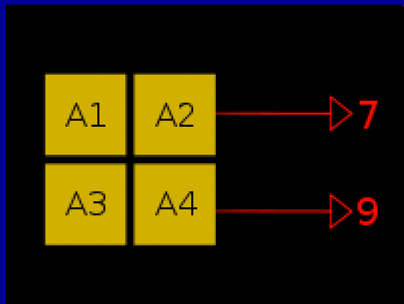
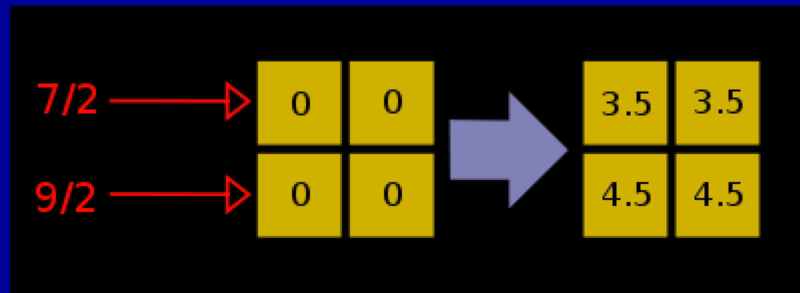
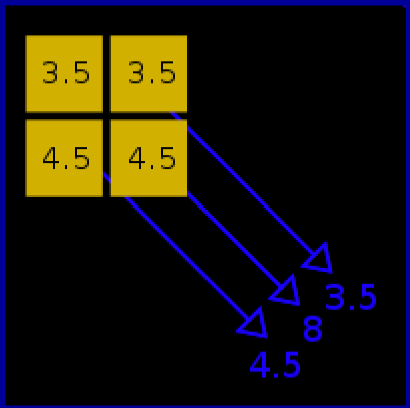
Iterative reconstruction



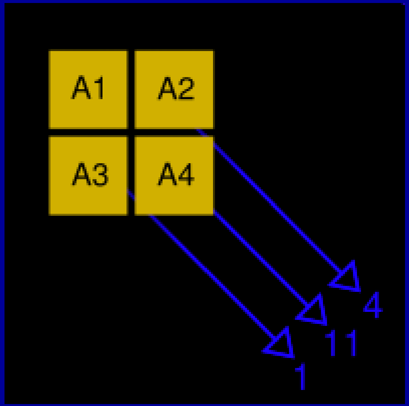
Actual projection, P1



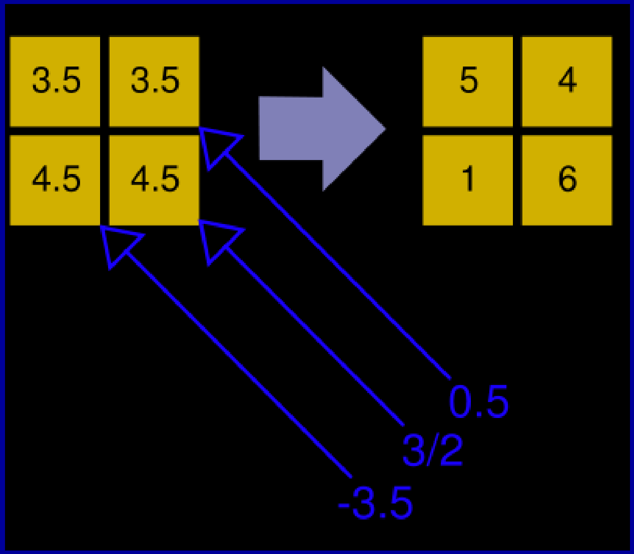
First estimate of image matrix.



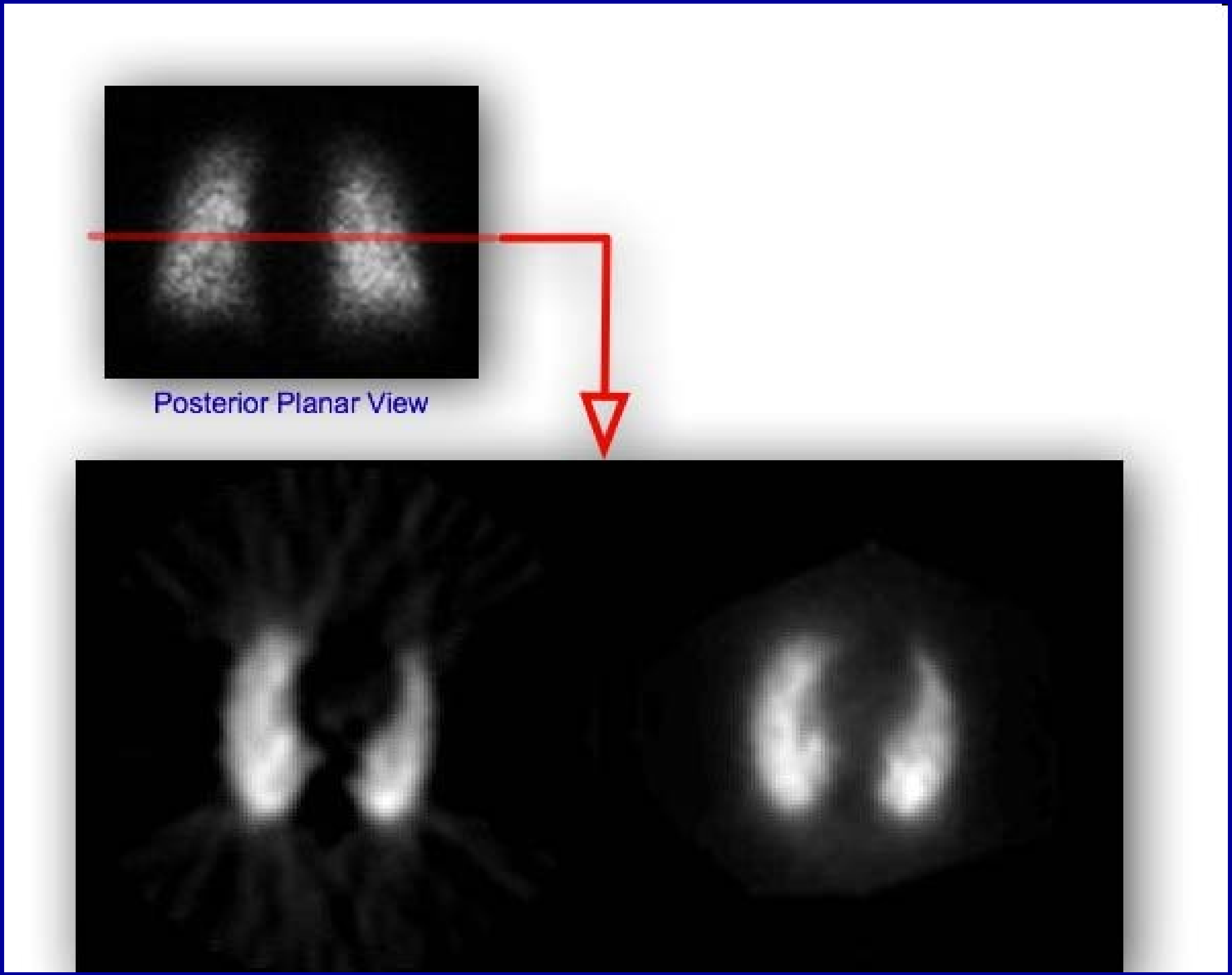
Estimate of projection, P2



Actual projection, P2



Second estimate of image matrix.



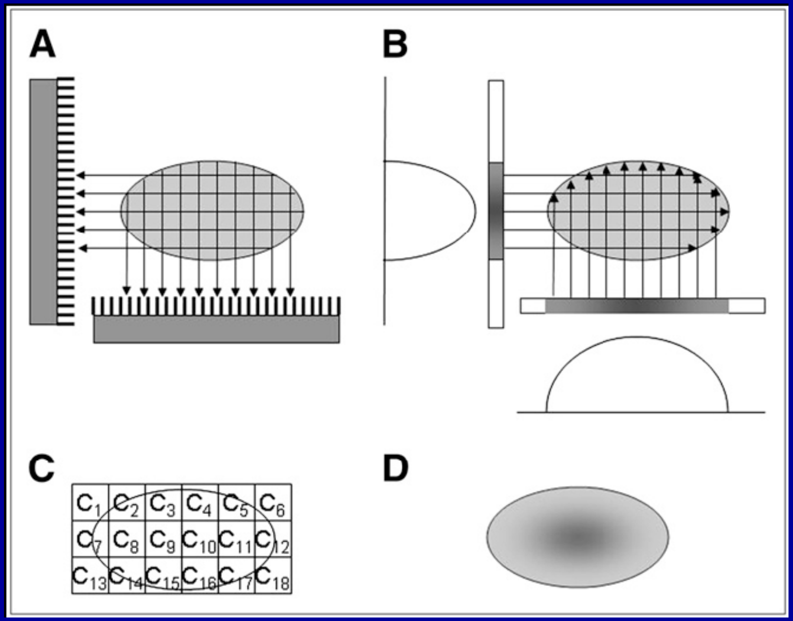
Slices reconstructed using Filtered Back Projection

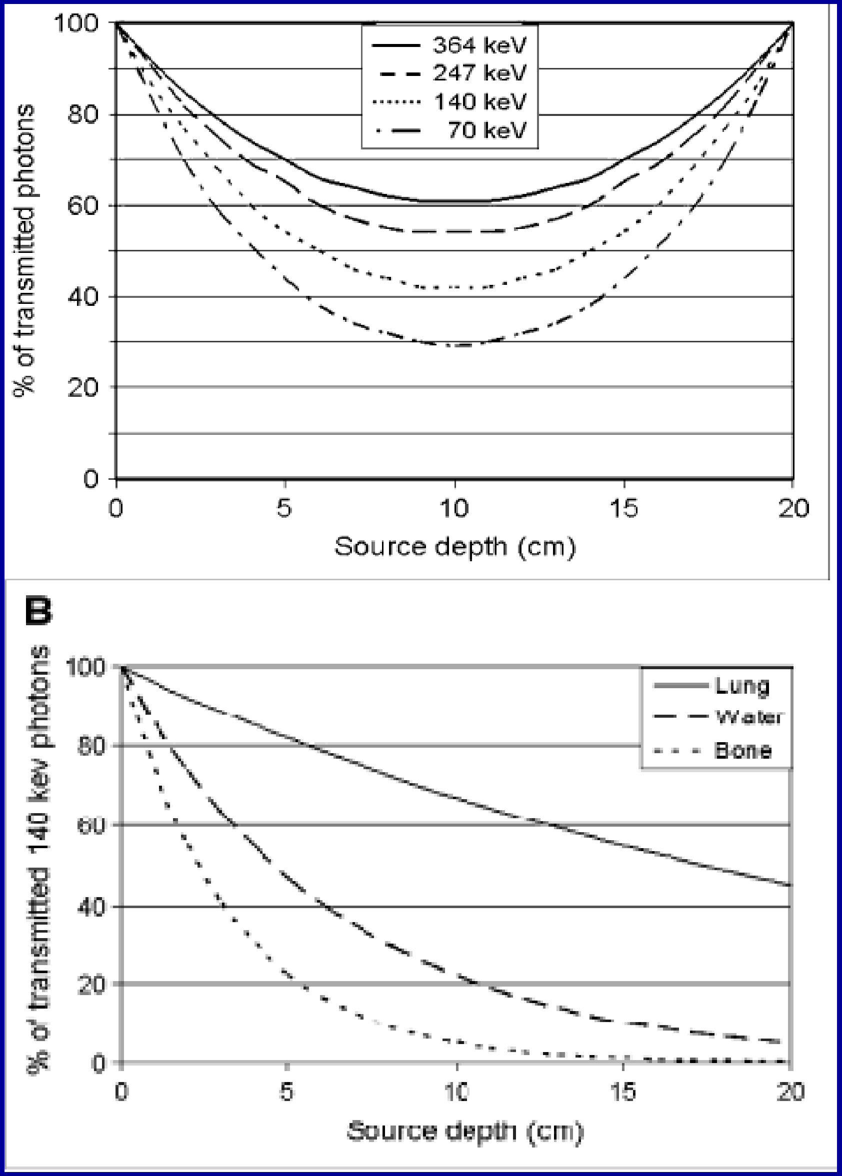
Slices reconstructed using Filtered Back Projection

SPECT degrading factors

- these images suffer from poor spatial resolution (typically >1 cm) and often lack anatomic landmarks
- Normal physiologic distributions must often be differentiated from regions of abnormal uptake
- Even though the source distribution is uniform, the reconstructed image shows an apparent decrease in activity

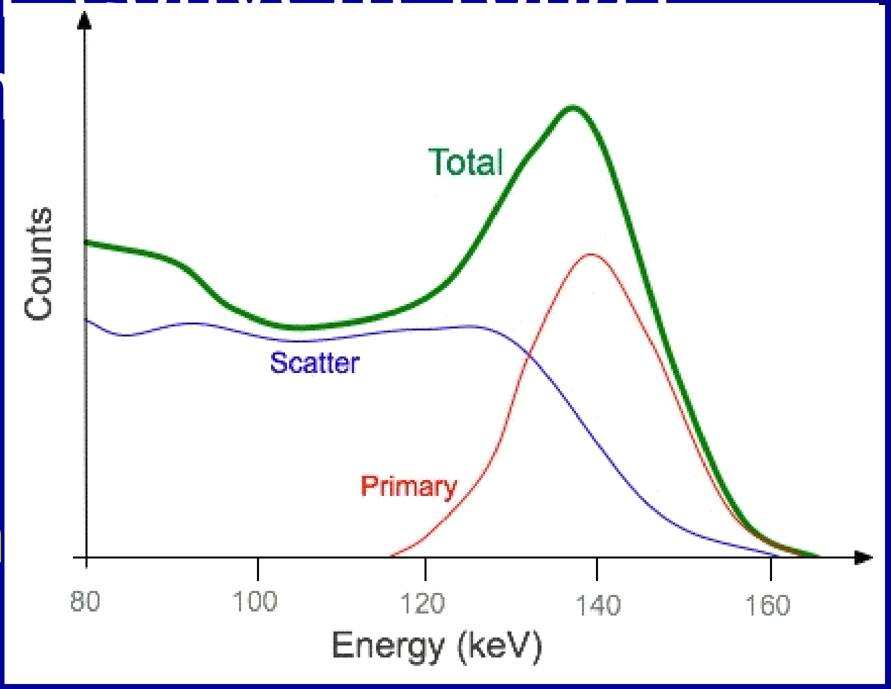
• Photon Attenuation





- The effects of attenuation are more intense at
  - lower energies
  - dense material

- The primary mechanism for attenuation in tissue throughout the diagnostic energy range in conventional nuclear medicin
  scattering
  - Loss of count
  - False backgroun



Why CT?

- CT provides high quality and high-spatial-resolution (~1 mm) images of cross-sectional anatomy.
- CT provides a significant portion of the anatomic images acquired in oncologic applications, not only for diagnosis and staging of disease but also for

Why SPECT/CT?

- The advantage of using CT data for attenuation correction of emission data
  - It provides a high photon flux that significantly reduces the statistical noise associated with the correction in comparison to other techniques (i.e., radionuclides used as transmission
  - The total imaging time is significantly reduced because of fast acquisition speed of CT scanners

- the anatomic images acquired with CT can be fused with the emission images to provide functional anatomic maps for accurate localization of radiopharmaceutical uptake

CT images are acquired as transmission maps with a high photon flux and are high-quality representations of tissue attenuation and thus can provide the basis for attenuation correction.

Thank you