Introduction to Emission Tomography

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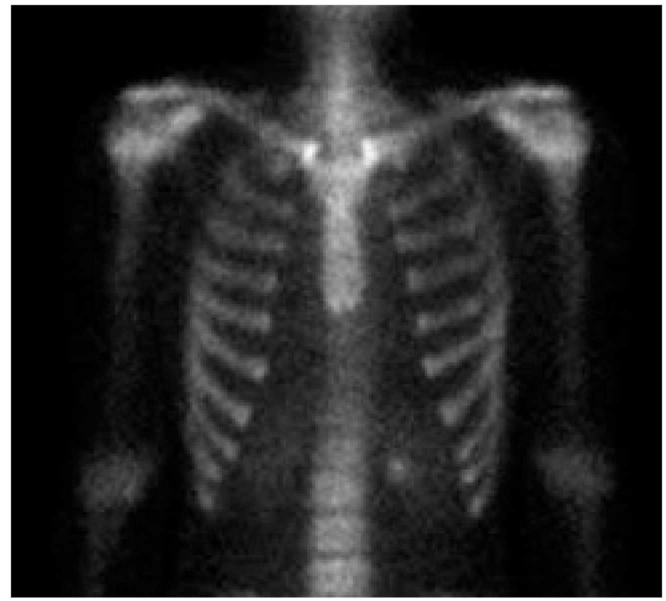
Gamma Camera Planar Imaging

Gamma Camera:

- collimator
- detector (crystal scintillator)
- data corrections
  - -linearity, energy, uniformity

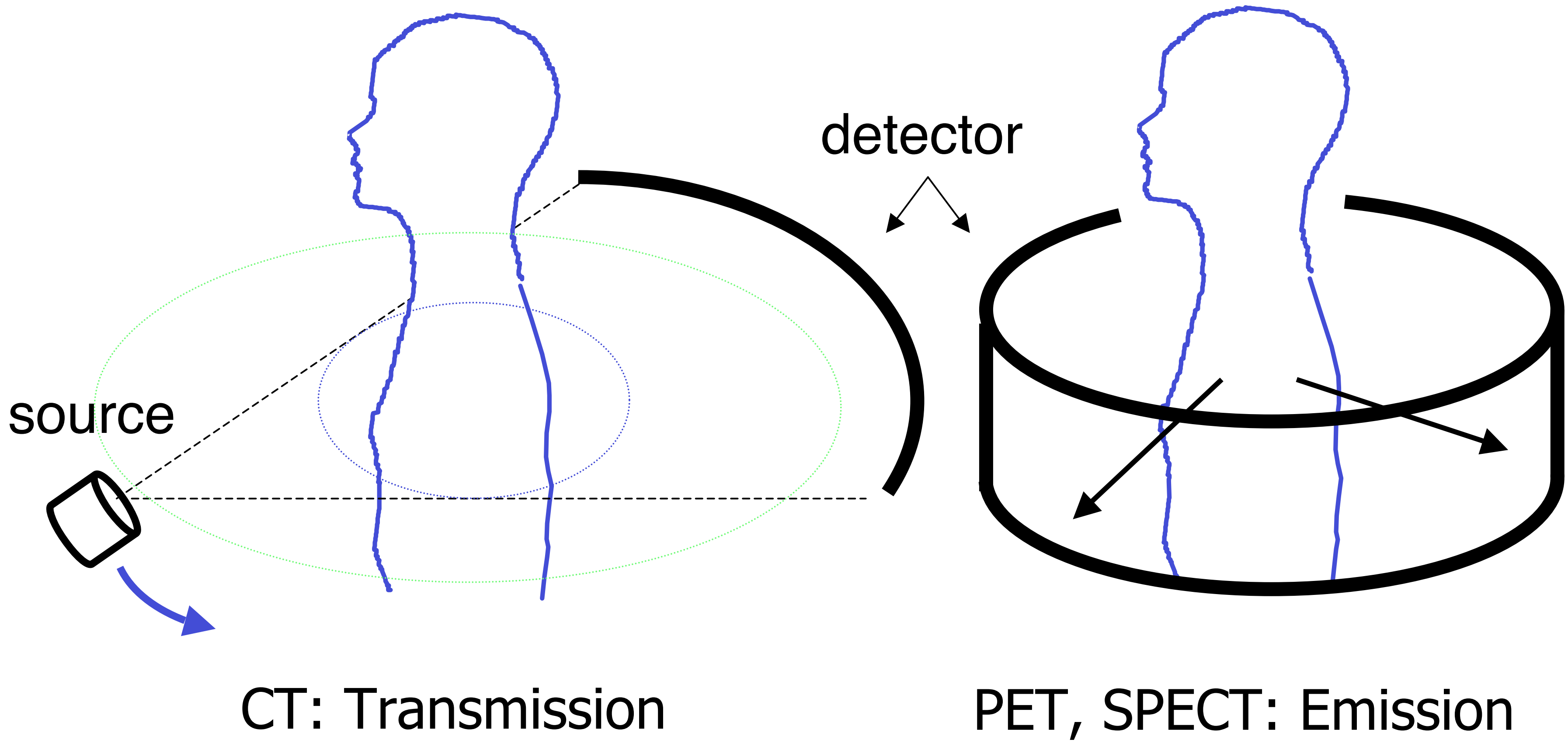


Planar Imaging:

- single projection view
- entire volume is projected onto one image plane
  - →structure overlap
- no image reconstruction is required

Emission and Transmission Tomography

'Tomo' + 'graphy' = Greek: 'slice' + 'picture'



- Projection views must surround patient.
- Eliminates overlap of structures in different planes.
- Requires image reconstruction.

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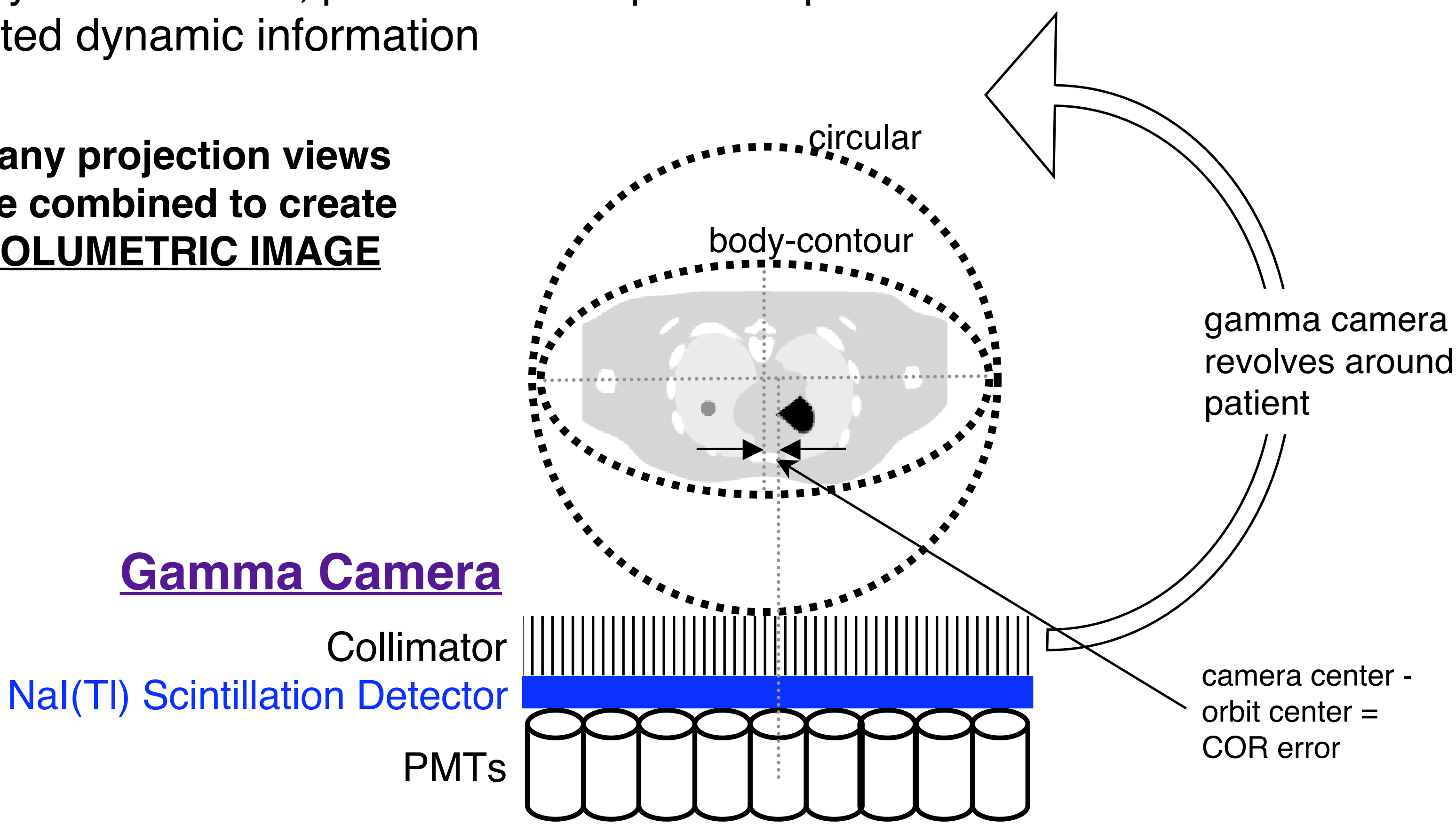
Single Photon Emission Computed Tomography (SPECT)

To acquire the necessary multiple projection views from different angles the gamma camera revolves around the patient in either (single, dual or triple head systems):

- circular orbit; simple, easy to implement
- body-contour orbit; preferred for improved spatial resolution
- limited dynamic information

Many projection views are combined to create <u>VOLUMETRIC IMAGE</u>



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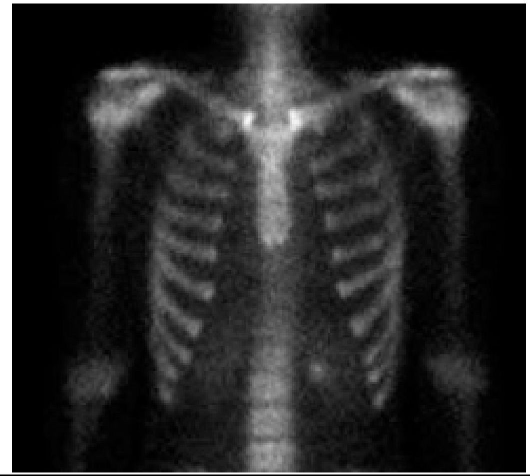
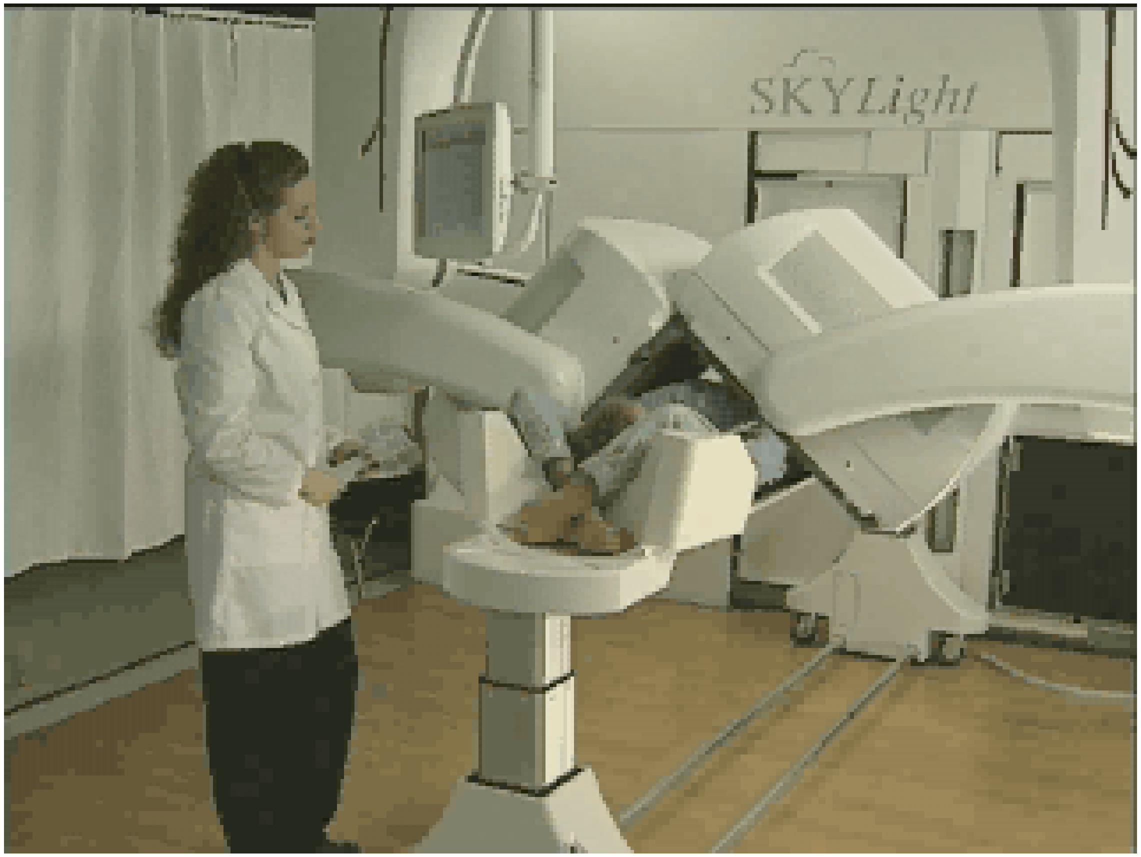
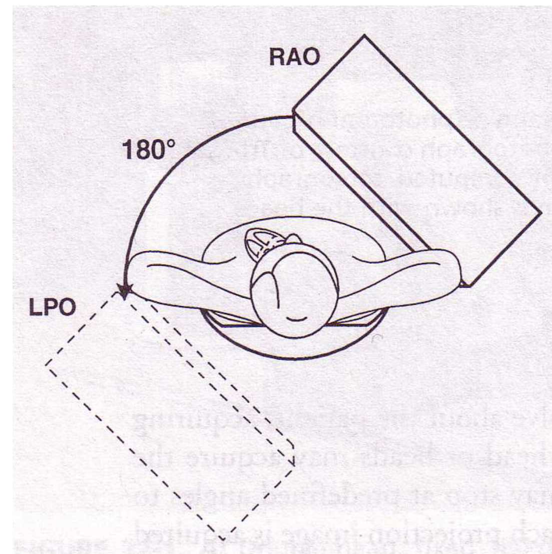
Gamma Camera Imaging (SPECT)



www.medical.philips.com

SPECT - 180 degree acquisition

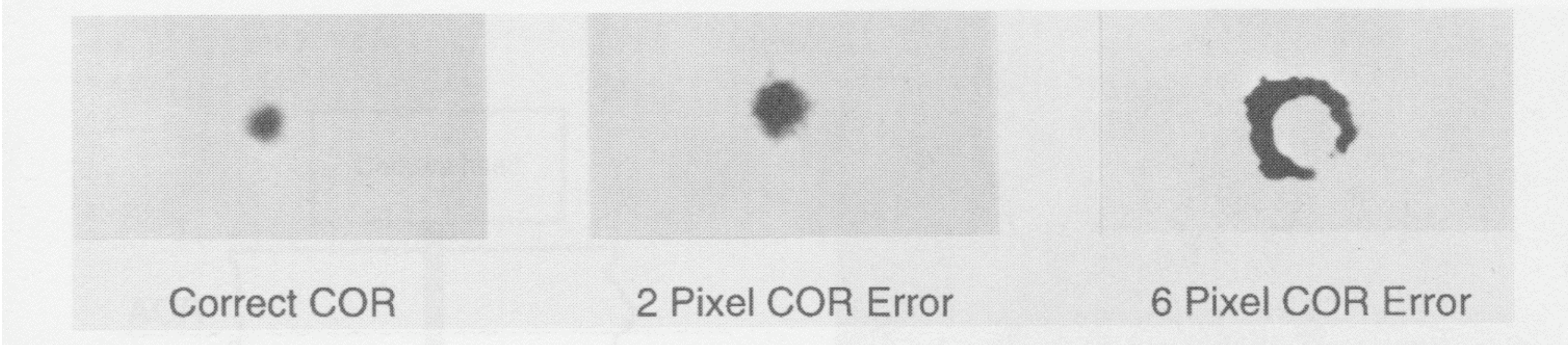




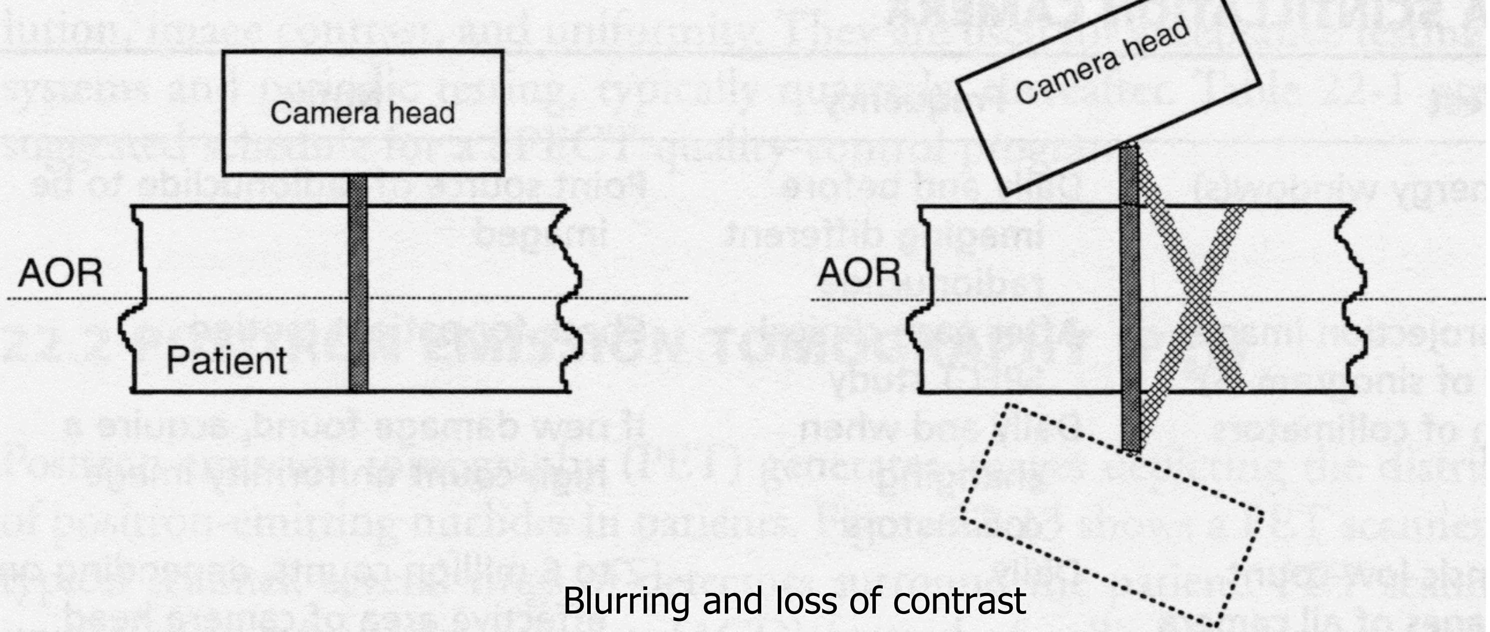
Cardiac -180 degree acquisition

Additional QA/QC Tests for SPECT

Center of Rotation



Camera Head Tilt

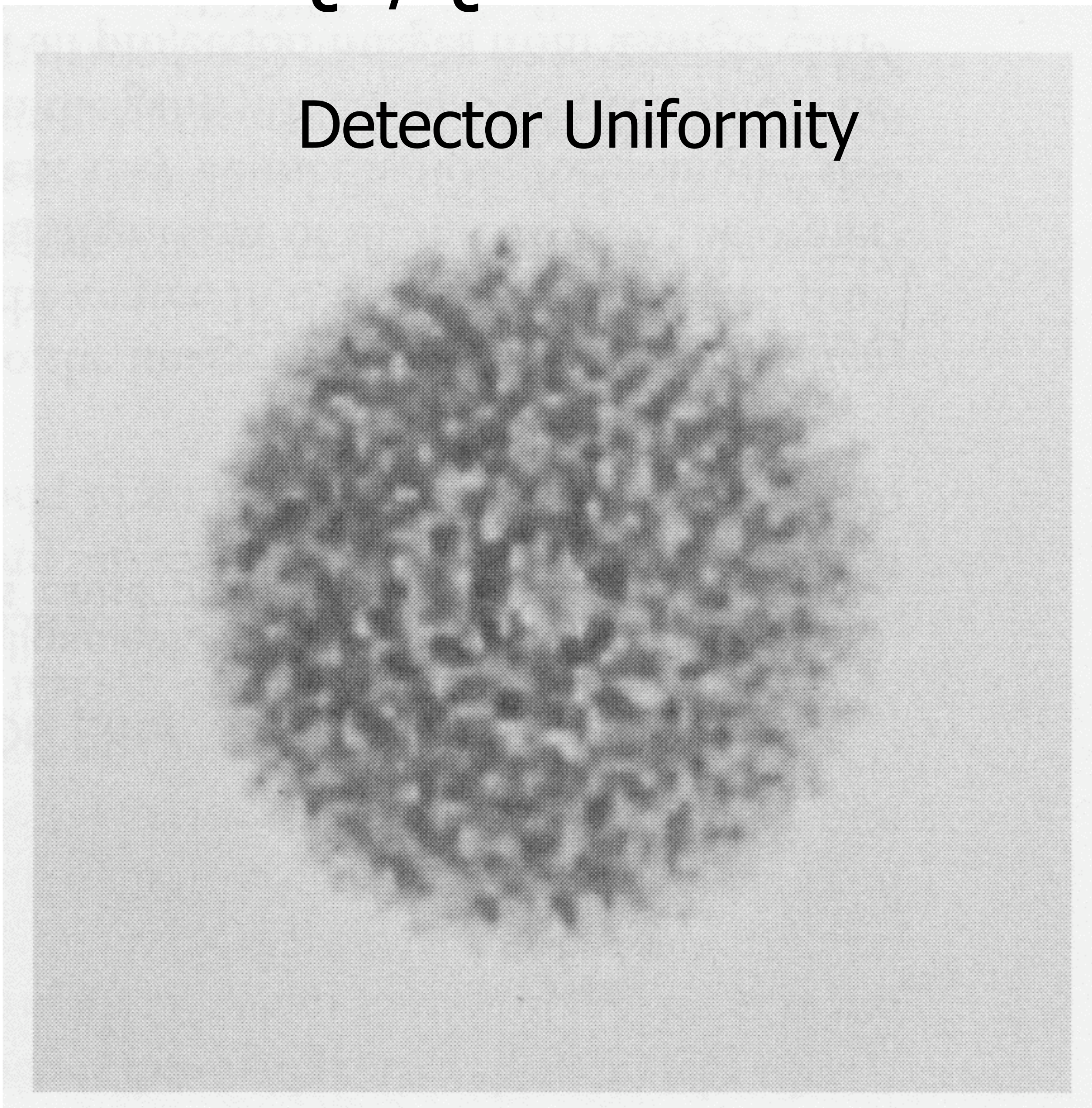


Blurring and loss of contrast

From: The Essential Physics of Medical Imaging (Bushberg, et al)

Additional QA/QC Tests for SPECT

Detector Uniformity

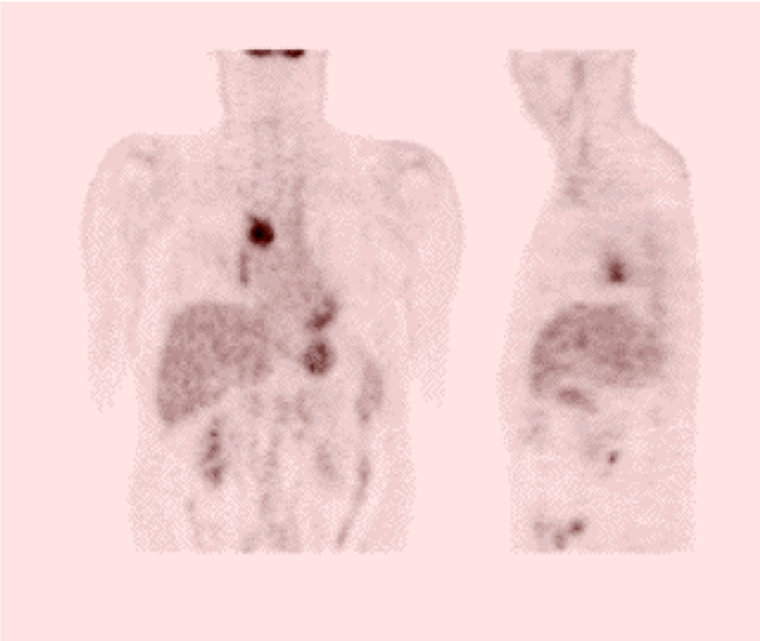


From: The Essential Physics of Medical Imaging (Bushberg, et al)

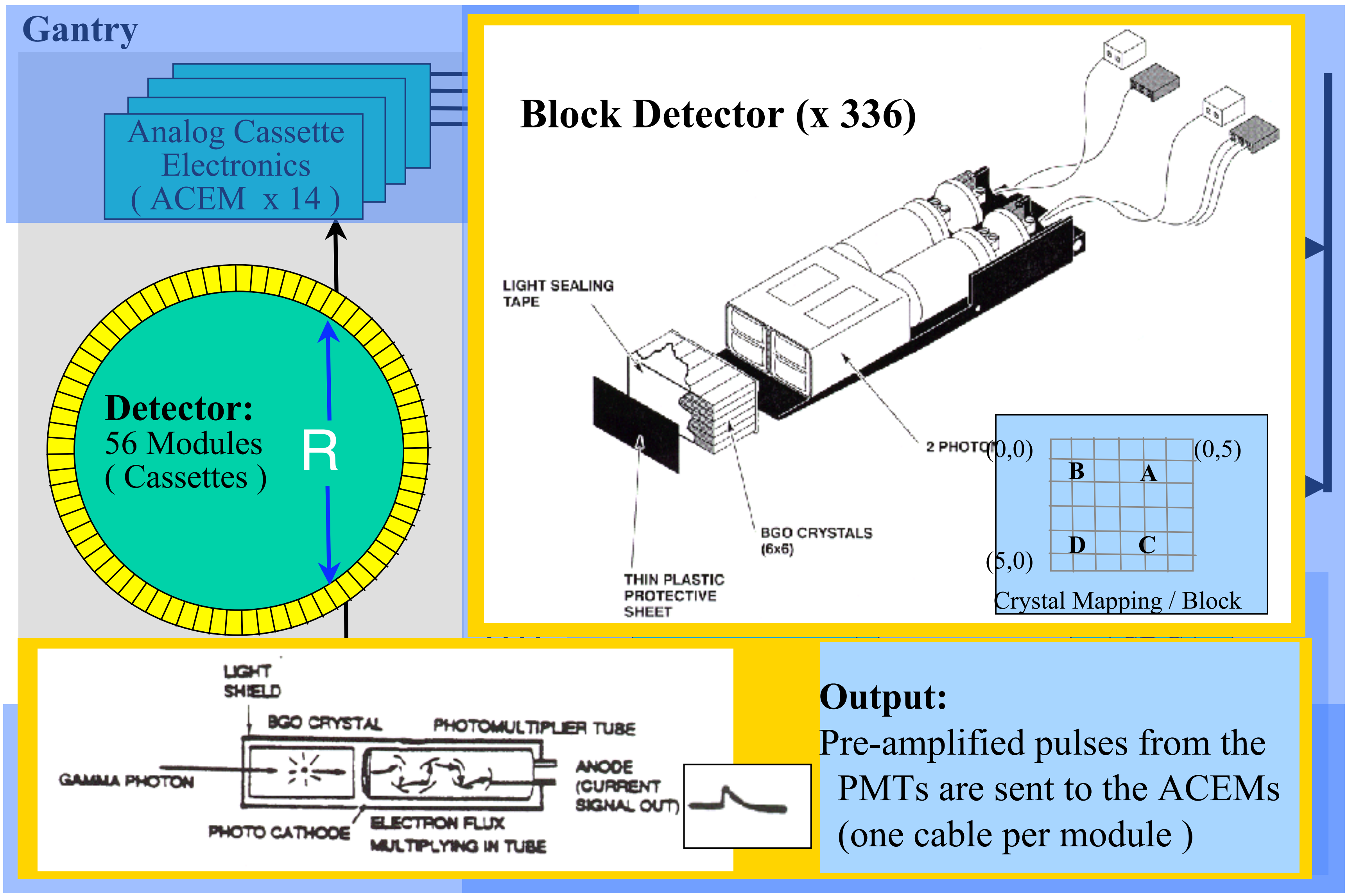
PET Scanner

All commercial PET scanners are now combined PET-CT systems

PET imaging is inherently tomographic
- i.e. no equivalent of the planar view in positron emission imaging
  (exception: positron emission mammography (PEM))

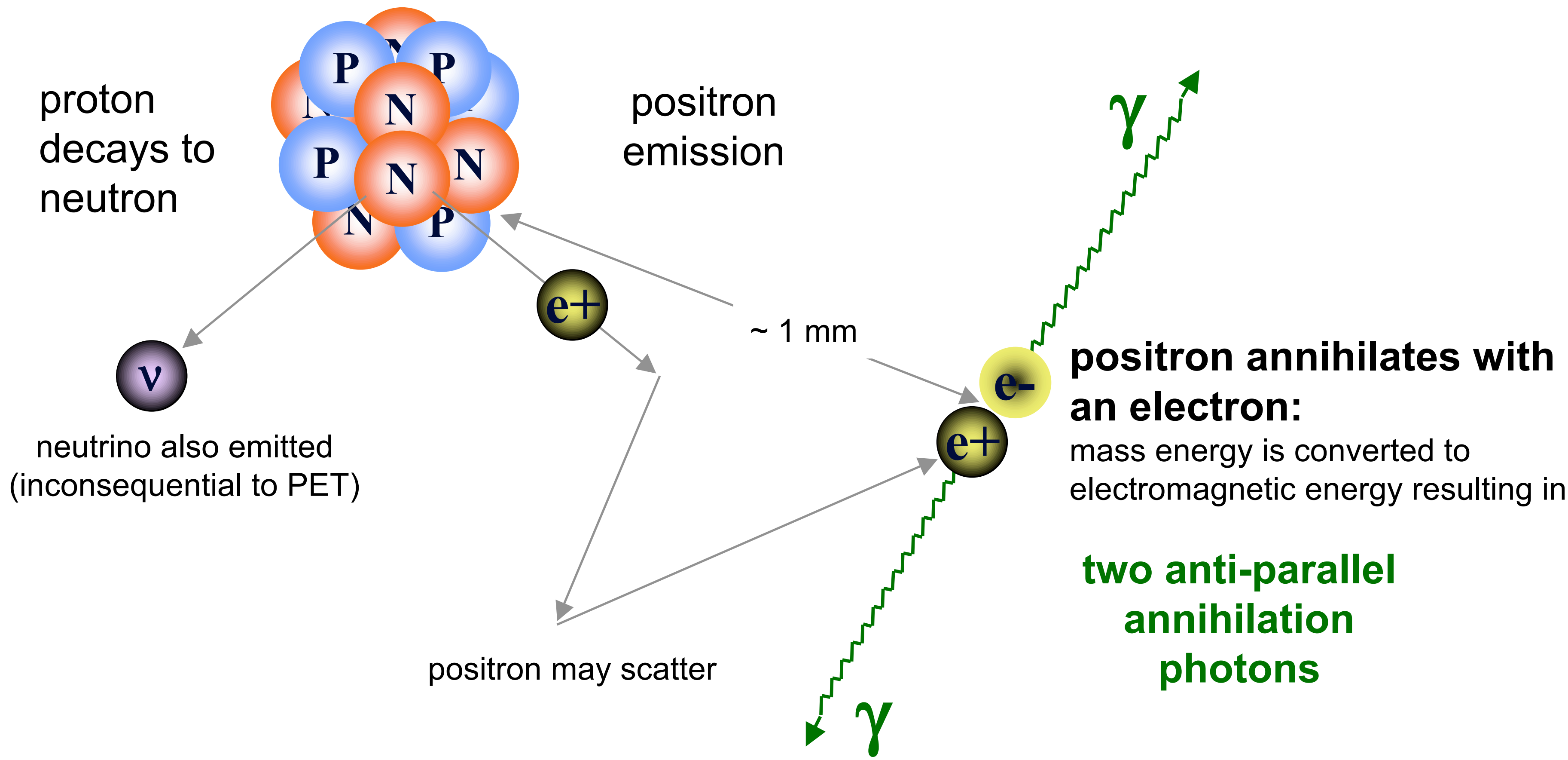


PET System Overview



Positron Annihilation

Parent nucleus:
unstable due to excessive P/N ratio
$(^{18}F, {}^{11}C, {}^{13}N, {}^{15}O, {}^{124}I) \longrightarrow (^{18}O, {}^{11}B, {}^{13}C, {}^{15}N, {}^{124}Te) + \beta^{+} + \nu$



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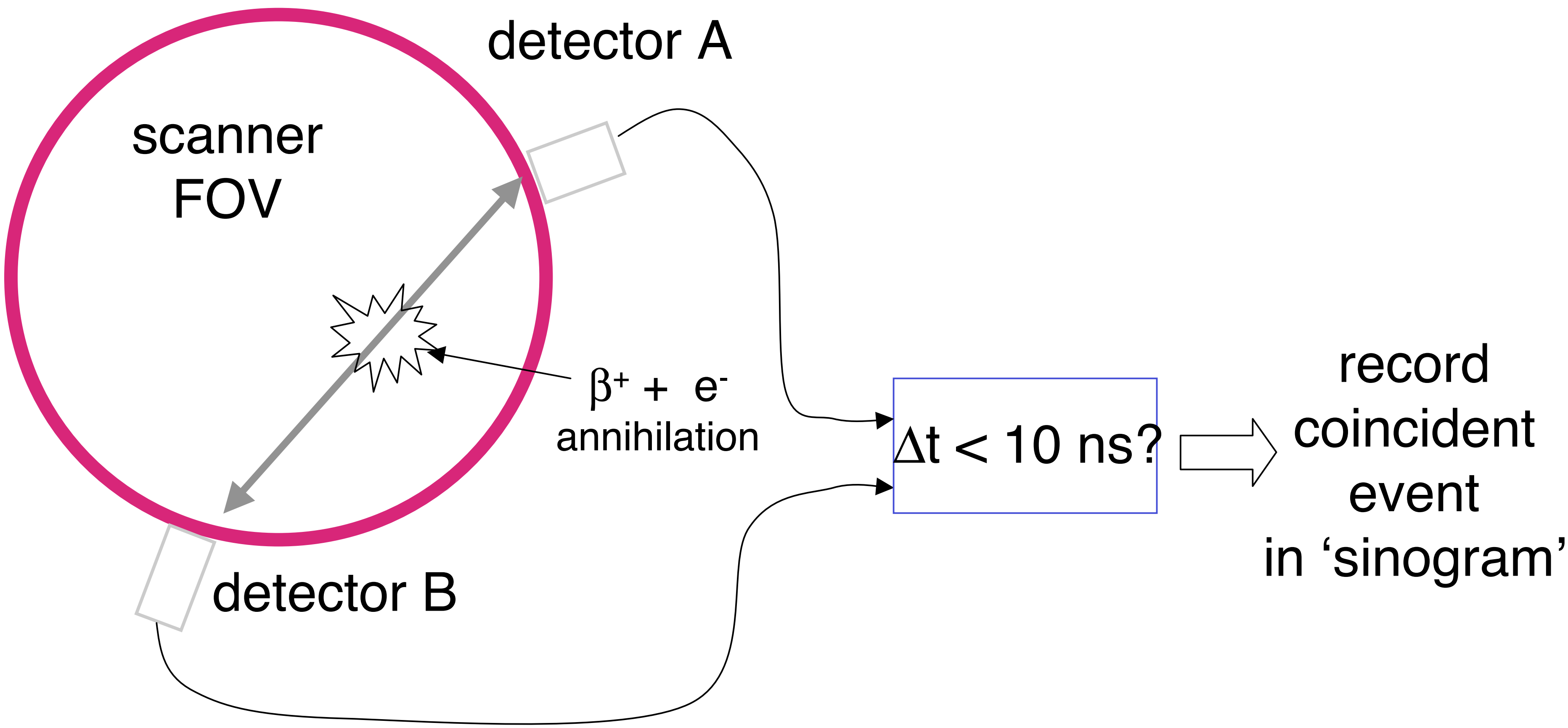
PET Coincidence Timing



- No need for collimation, thus much higher sensitivity than SPECT
- Also correction for attenuation is easier.
- Simpler chemistry, but you need a cyclotron

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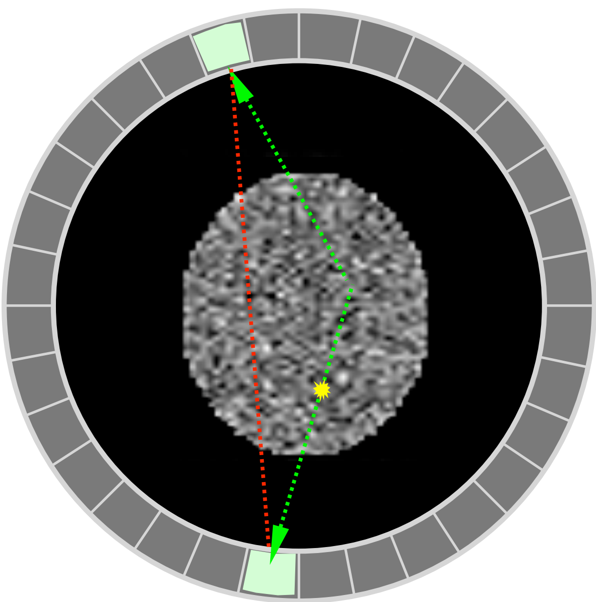
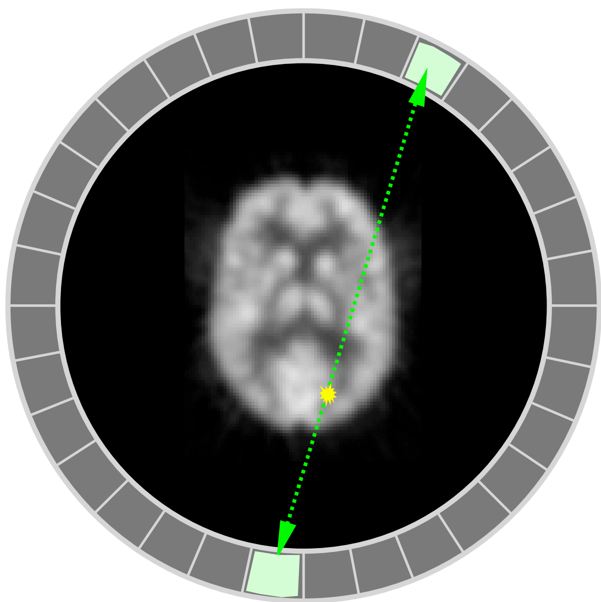
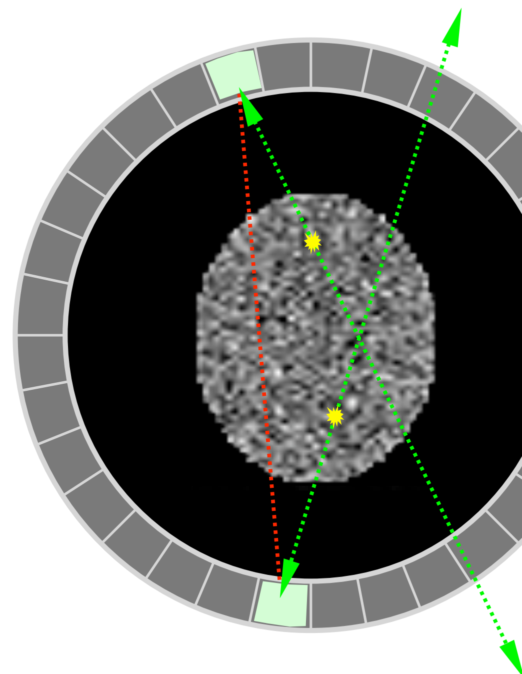
Coincidence Detection

PET detectors seek simultaneous gamma ray absorptions
("simultaneous" → within <10 ns)







Scattered coincidence:
one or both photons change direction from a scatter before detection
(low frequency bias)

True coincidence:
anti-parallel photons travel directly to and are absorbed by detectors

Random coincidence:
photons from different nuclear decays are detected simultaneously
(nearly uniform bias)

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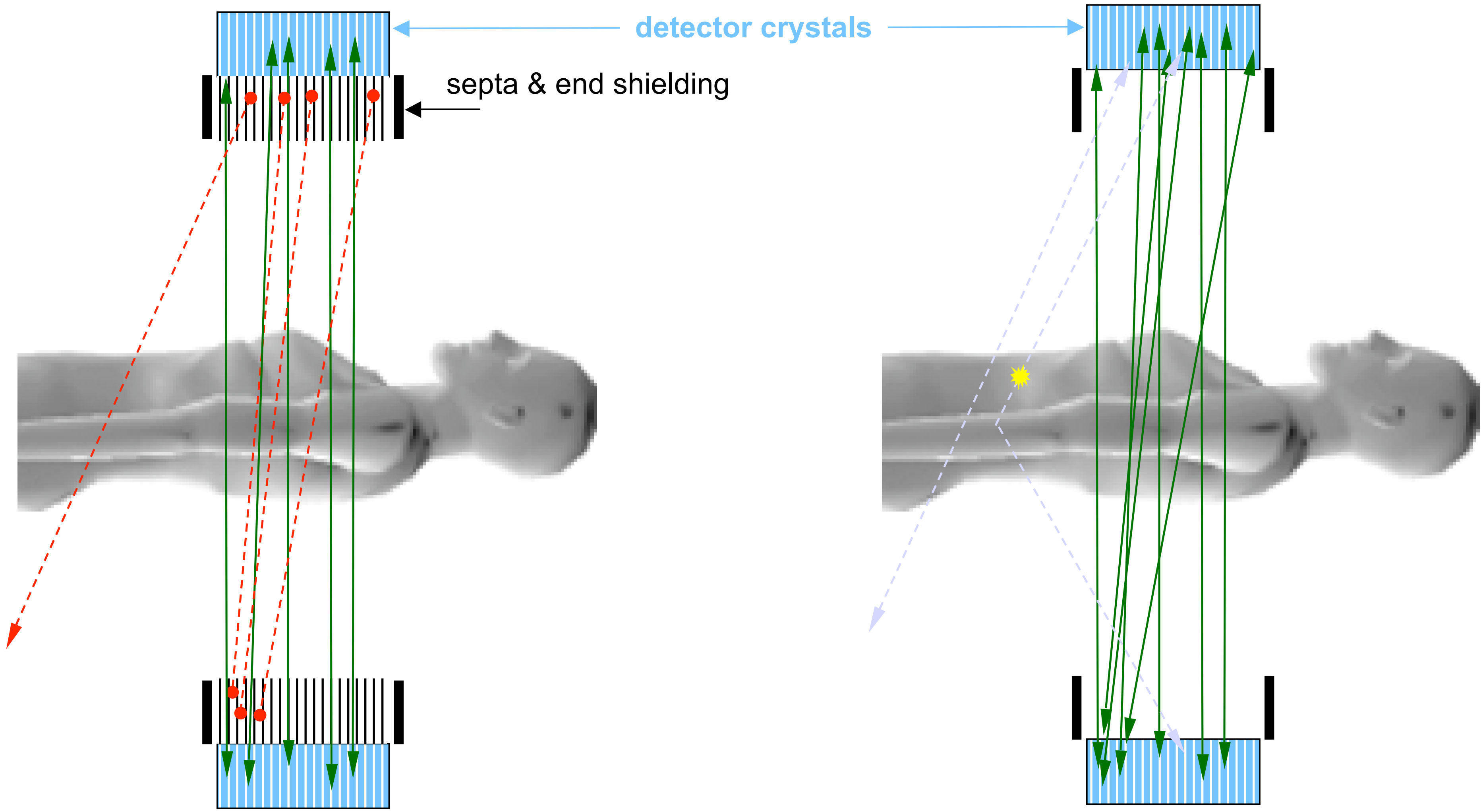
2D versus 3D PET

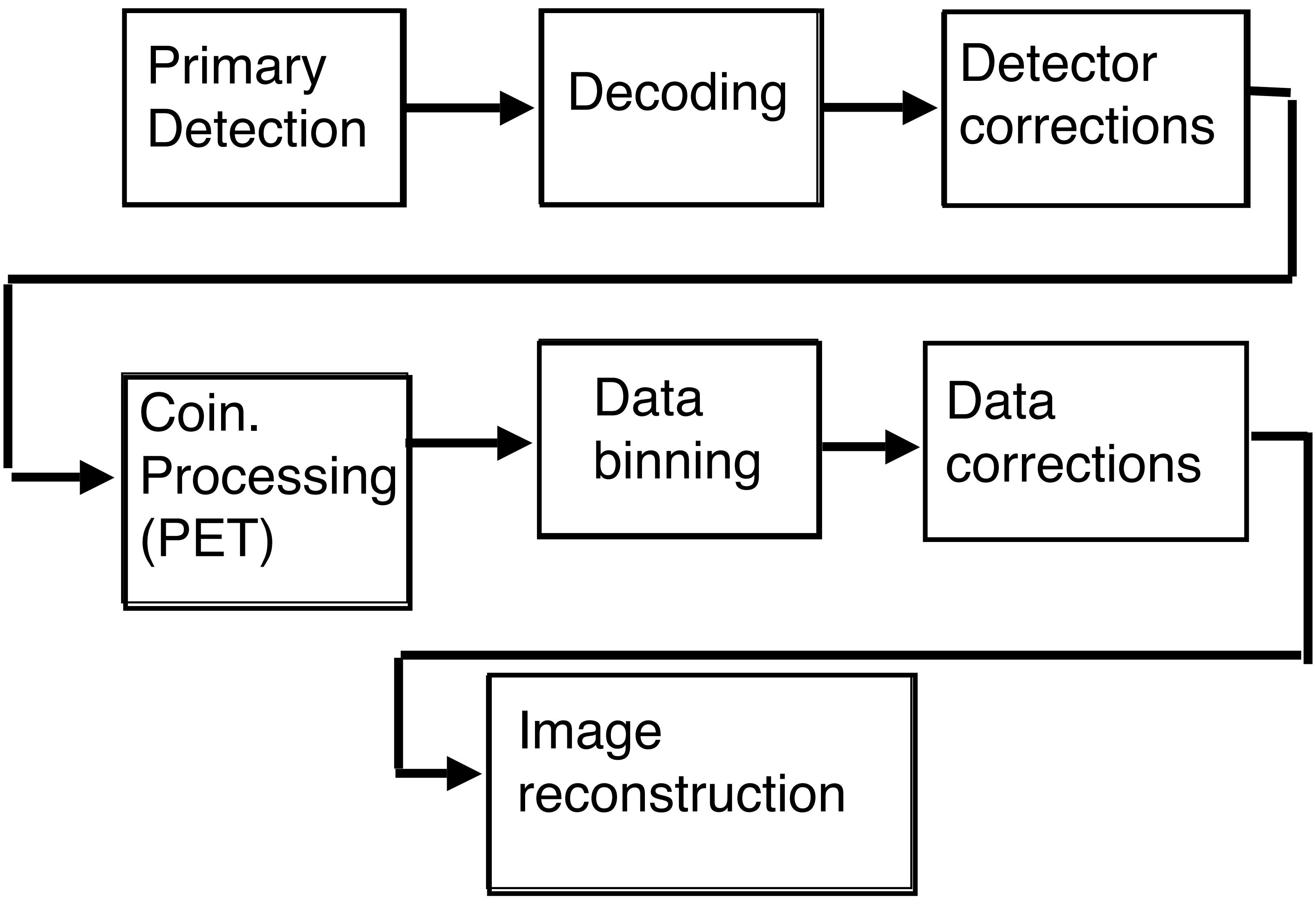
Shielding is used in the form of septa that separate axial slices in 2D PET
- reduces scattered and random events (also reduces trues!)



2D PET uses axial septa

3D PET uses no septa

The basic steps to make a tomographic image



Data corrections for emission tomography

- Attenuation
- Scatter
- Randoms (PET only)
- Normalization
- Dead time
- Partial volume (resolution)

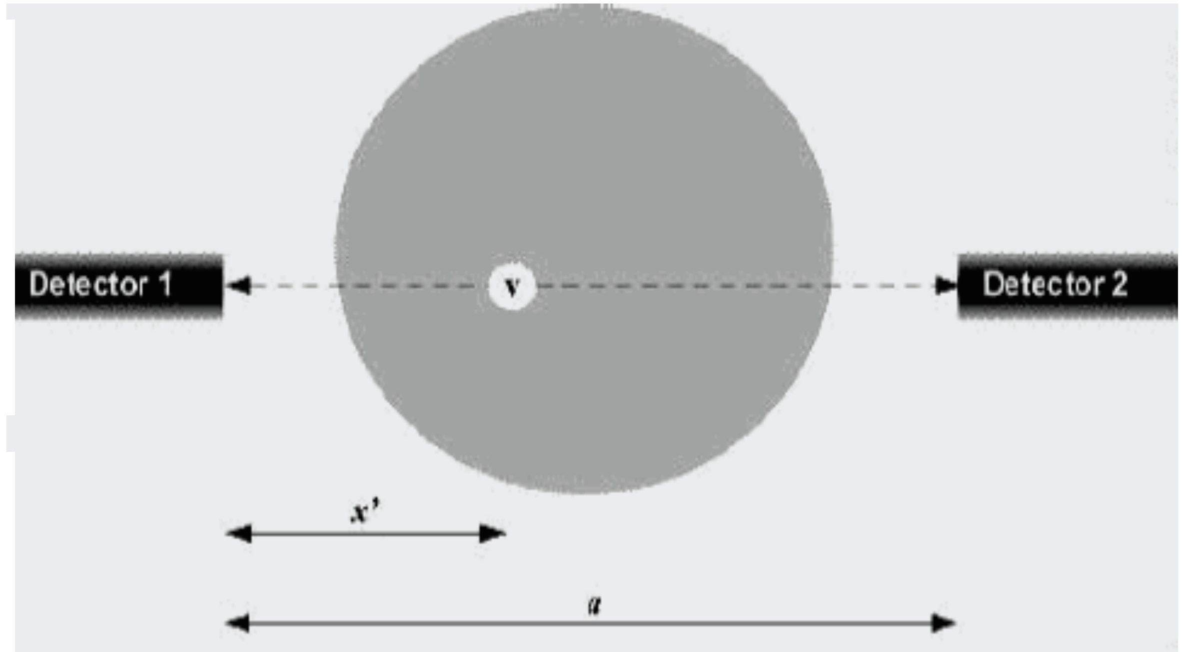
Attenuation

True of False

Most attenuation in PET and SPECT is caused by photoelectric absorption of photons within the patient?

PET: Attenuation independent of depth

$$P_1 = e^{-\int_0^{x'} \mu(x)dx}$$

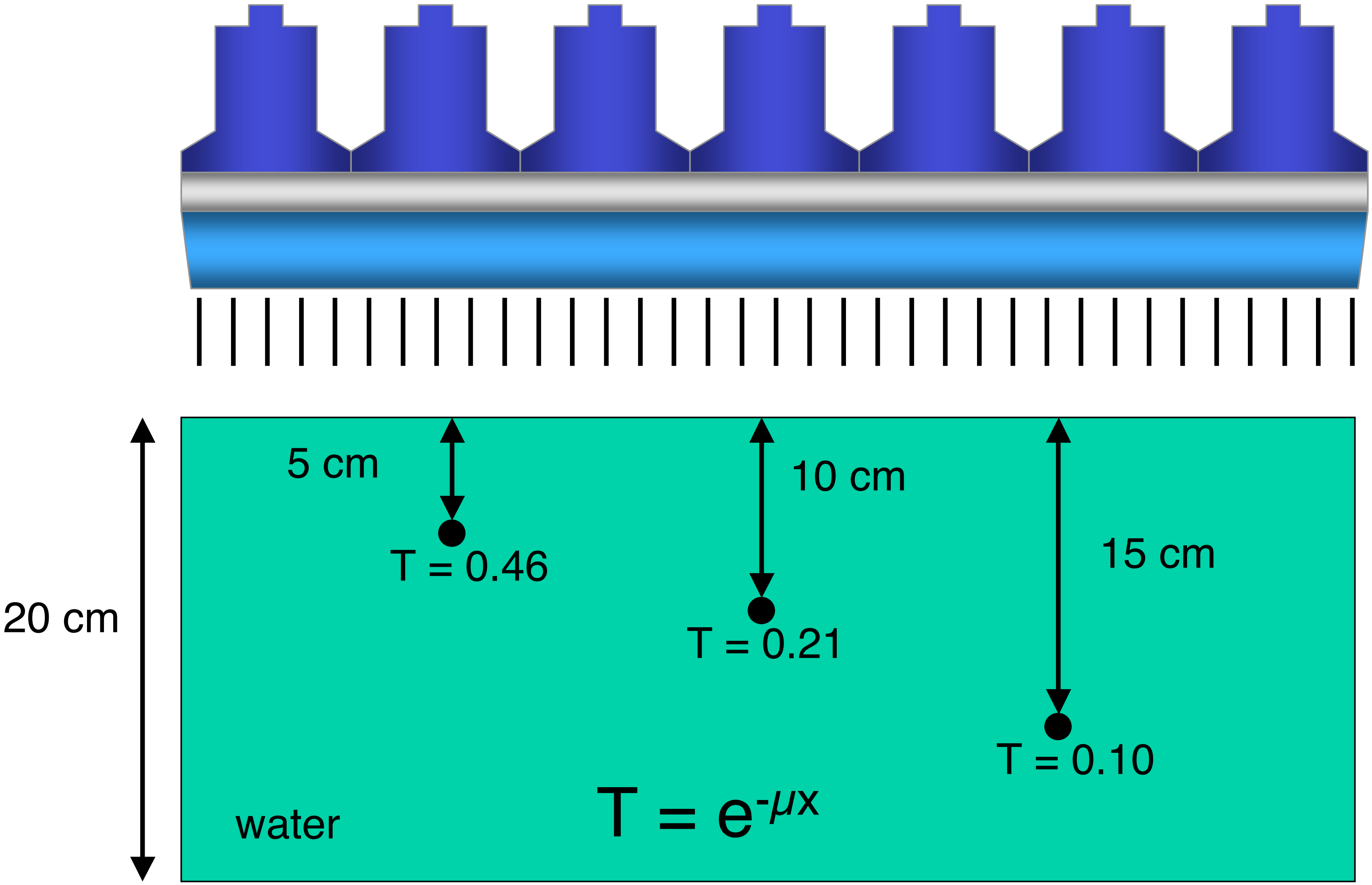


$$P_2 = e^{-\int_{x'}^{a} \mu(x)dx}$$

$$P_C = P_1 P_2 = e^{-\int_0^{a} \mu(x)dx}$$

Attenuation is the same for any point along a given LOR
Simple multiplicative attenuation correction

SPECT: Attenuation is depth dependent



μ =0.155 cm^{-1} for 140 keV γ's in water

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Attenuation Correction: SPECT

Chang attenuation correction
(approximate, image based approach)

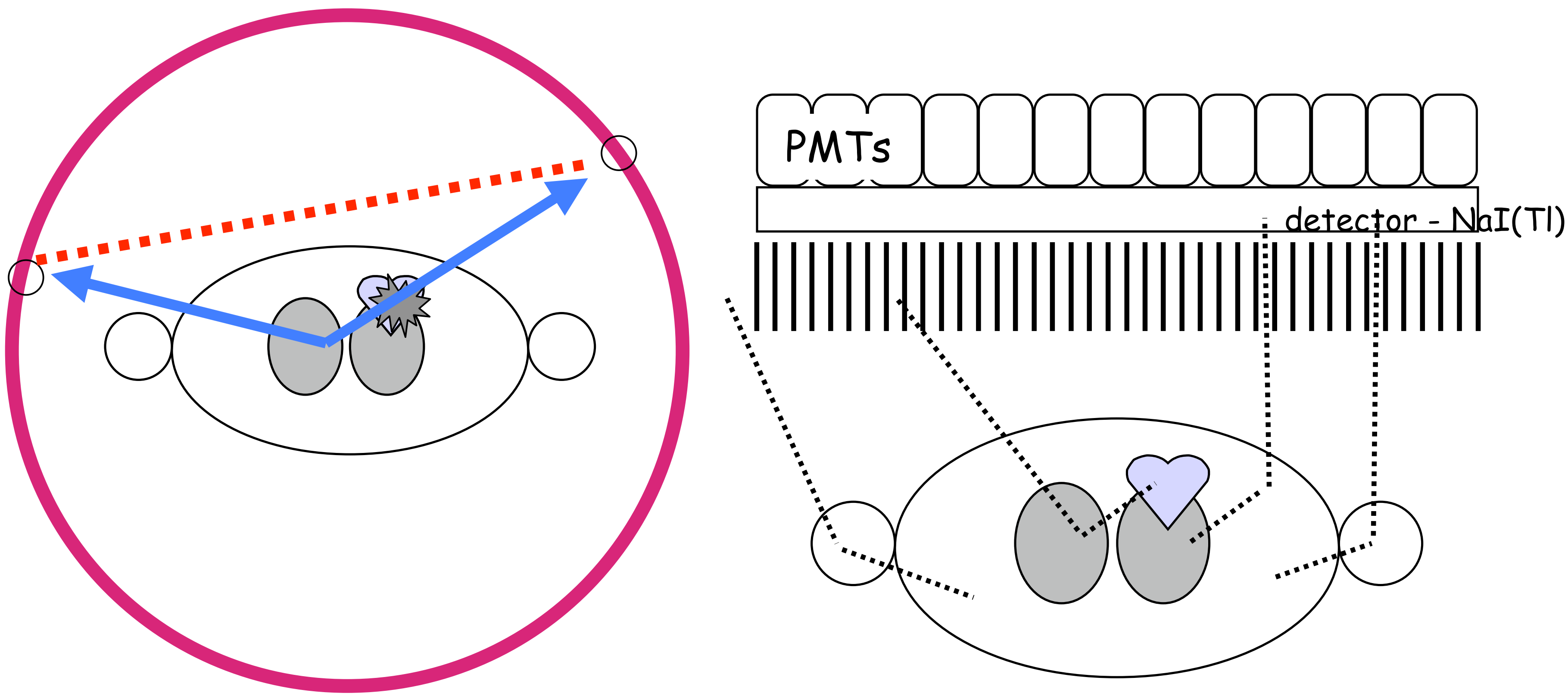
Attenuation included in system model of
iterative image reconstruction
(more accurate: requires attenuation
image and computationally expensive)

Transmission scanning

In past used external radiation sources

CT is preferred for both PET and SPECT

Scatter: PET vs. SPECT



PET scatter

SPECT scatter

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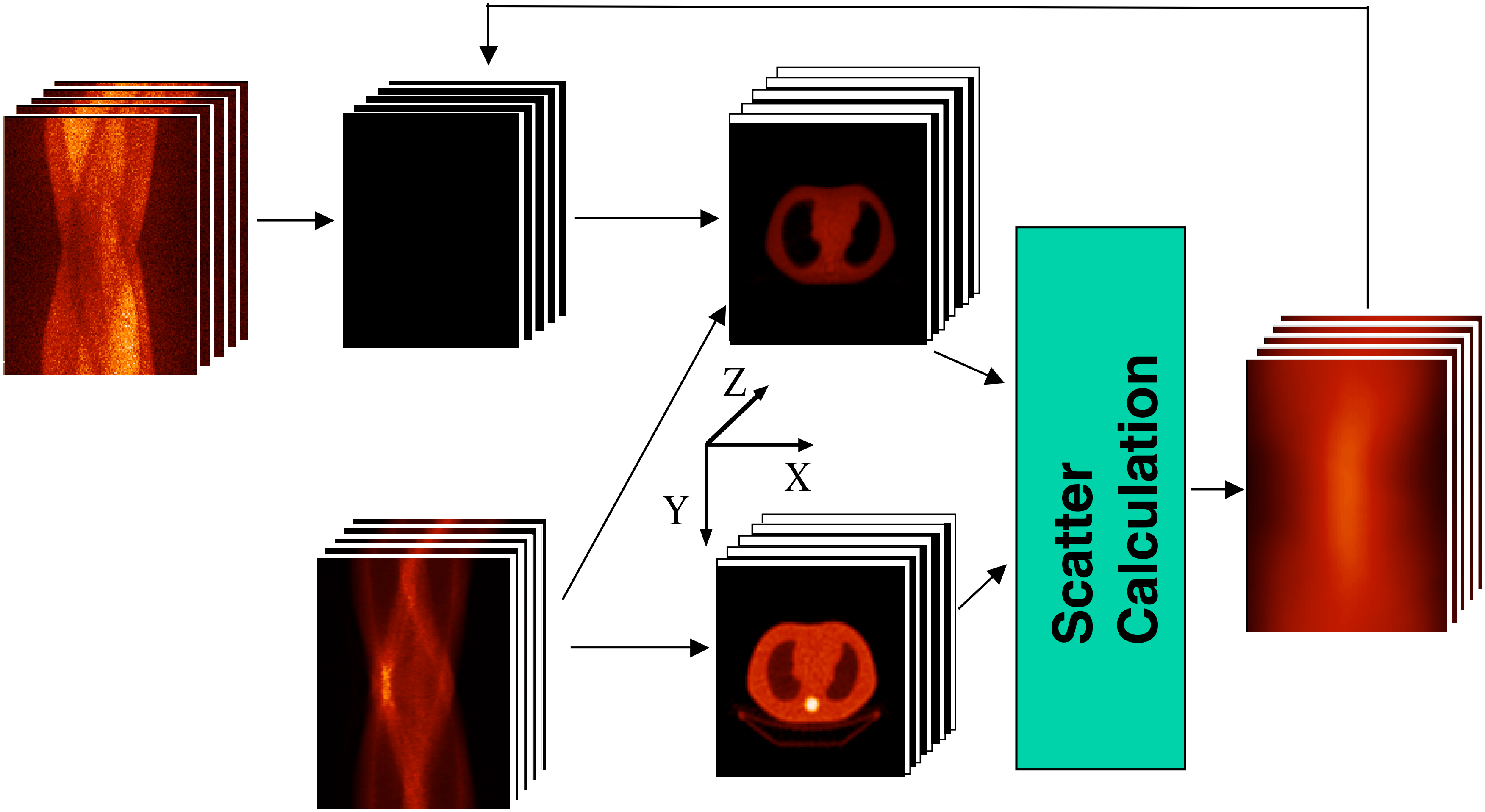
Scatter Fractions

Whole body PET: equal or > trues

^{99m}Tc cardiac: ~40% trues

Model based scatter correction algorithm



1. Estimate single scatters based on current emission & xmission images
2. Estimate multiple scatters with convolution model
3. Subtract scatter from emission sinograms and iterate
4. Scale final estimates to force mean value of pixels outside object to 0

From: Lewellen fall 2005

SPECT Scatter Corrections

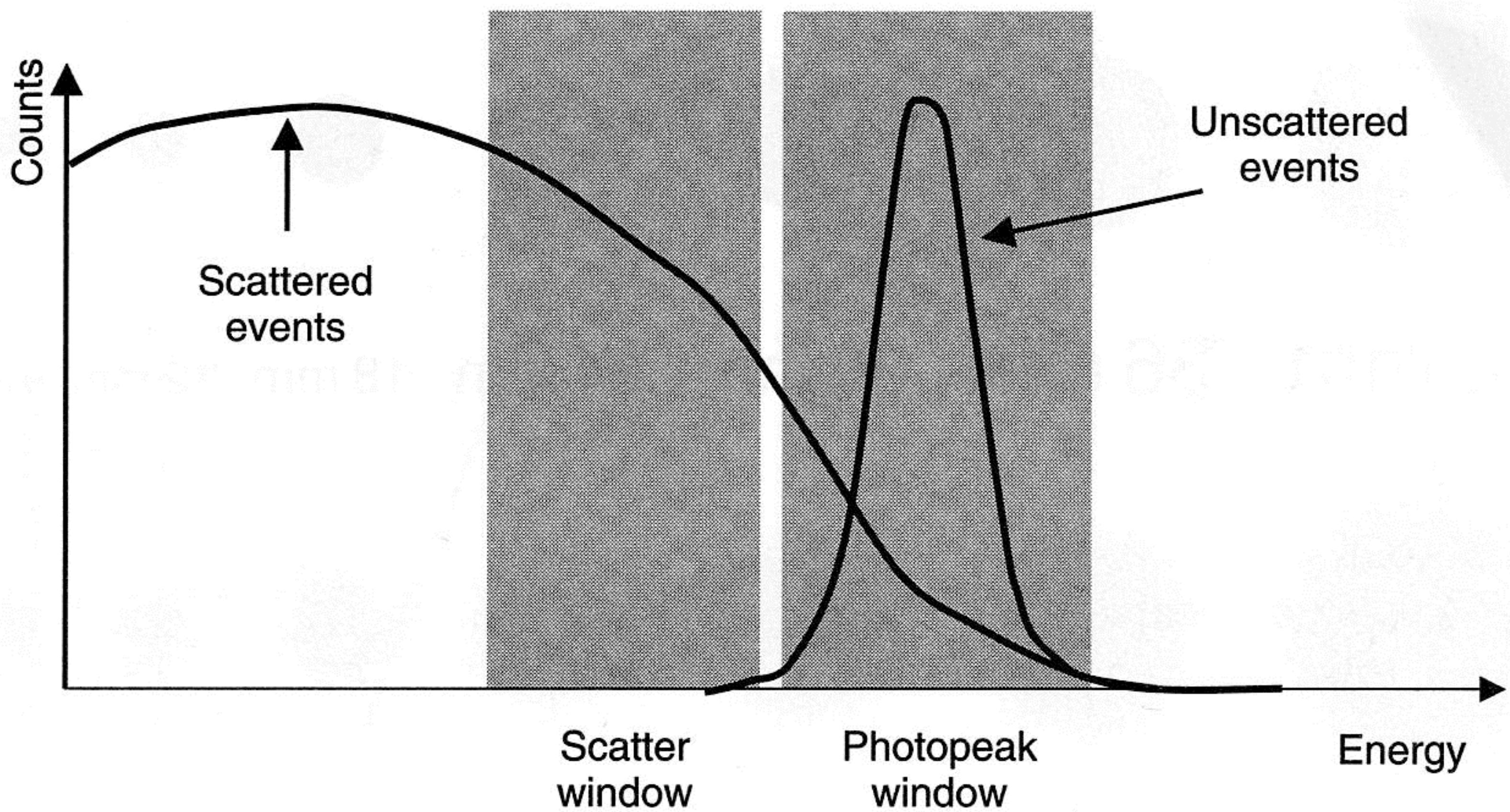
Use broad beam versus narrow-beam attenuation coefficient
- ~0.12 cm^{-1} versus 0.155 cm^{-1} water at 140 keV
- simple but not very accurate

Deconvolve measured scatter projection profiles from data
- scatter projection profiles object dependent
- usually measured for only one or two sized objects

Scatter correction included in iterative image reconstruction
- does not account for scatter from outside the FOV

Energy-based scatter correction technique

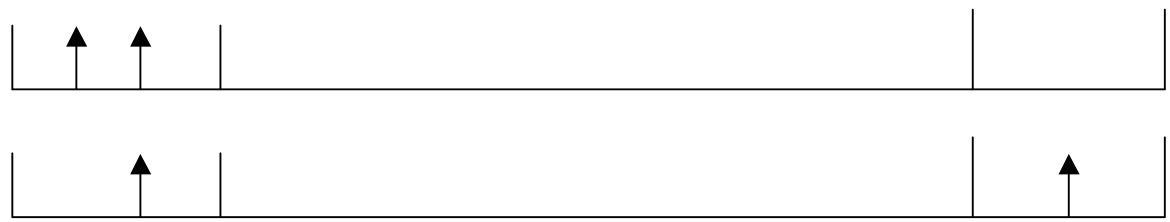
Energy-based Scatter Correction



Scatter counts in photopeak window estimated as a fraction of counts in the scatter window.

Randoms Correction

Delayed timing window - real time subraction

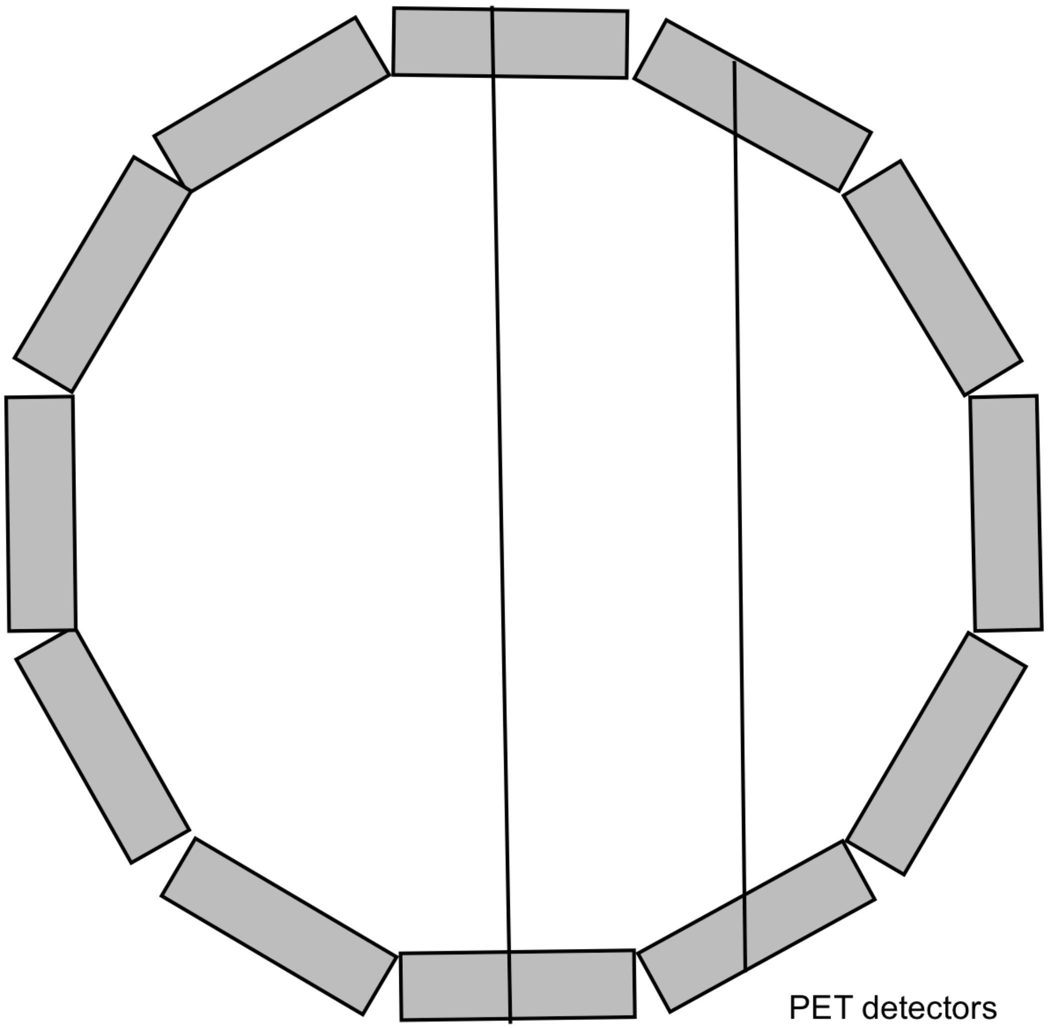


Delayed timing window - smoothed

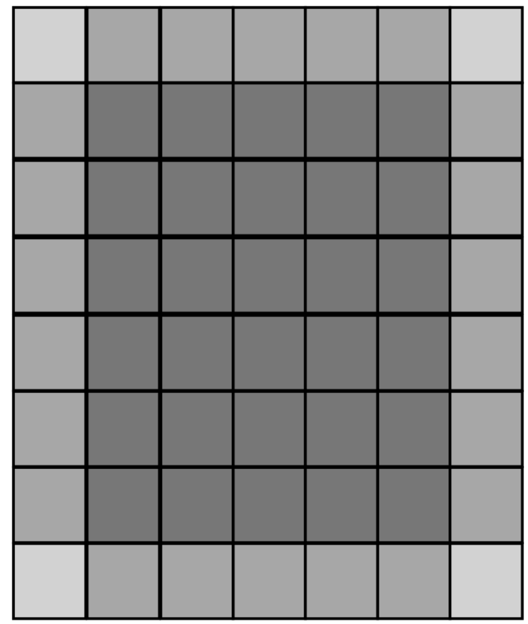
Model based techniques:
Randoms based on singles information

$$R_{LOR} = 2\tau * S_1 * S_2$$

Detector Normalization (PET)



Geometric efficiency



Crystal sensitivities

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Detector Normalization (SPECT)

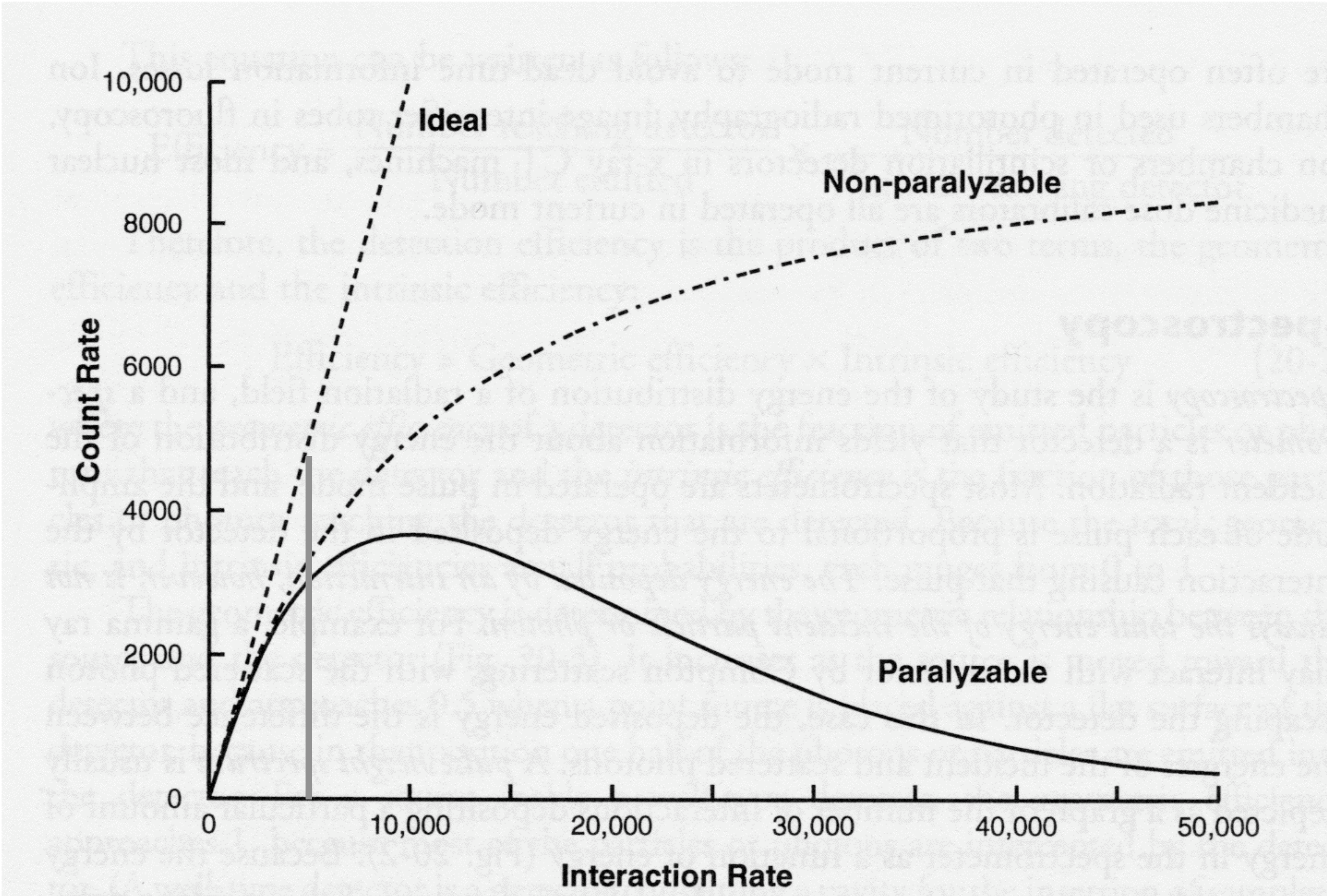
Detector uniformity

Collimator uniformity

Different normalization files for different energies

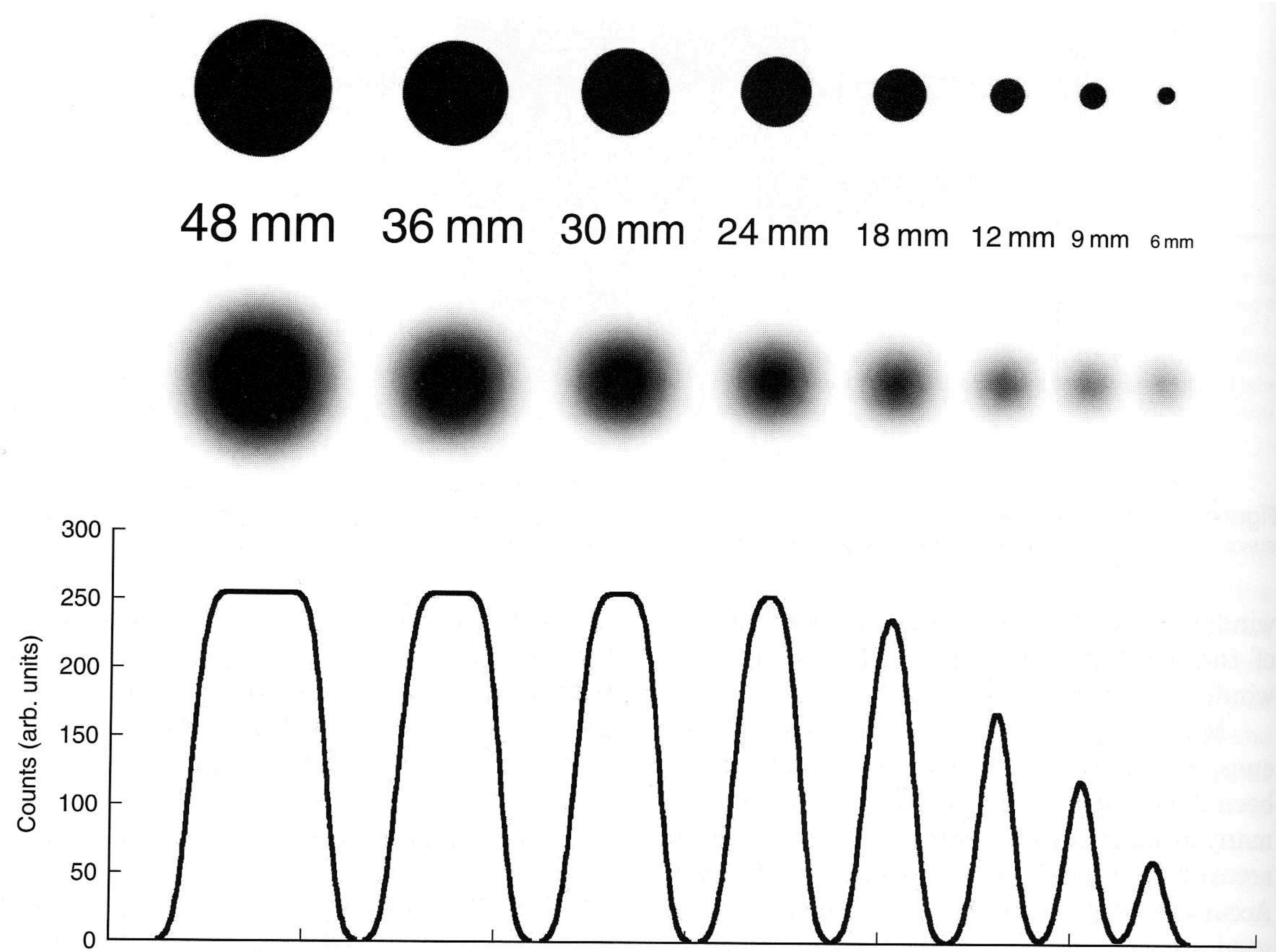
SPECT requires higher quality normalization files

Deadtime



From: The Essential Physics of Medical Imaging (Bushberg, et al)

Partial Volume Effect

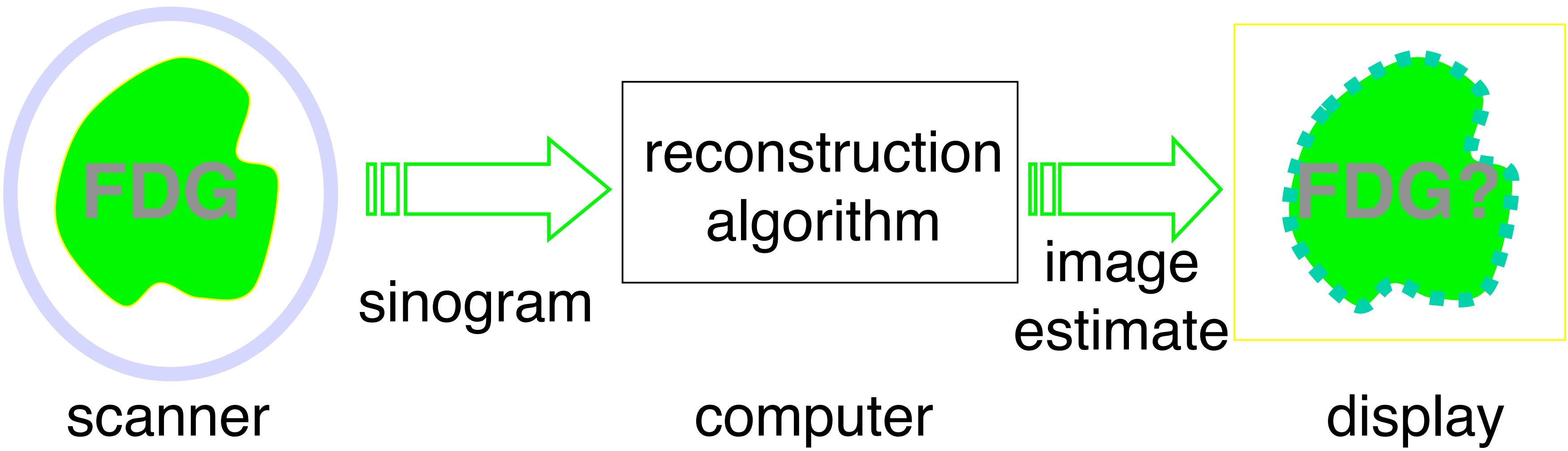


Object must be > 2*FWHM spatial resolution to avoid partial volume effect.
Images from a SPECT system with in-plane resolution of 12 mm FWHM.

From: Physics in Nuclear Medicine (Cherry, Sorenson and Phelps)

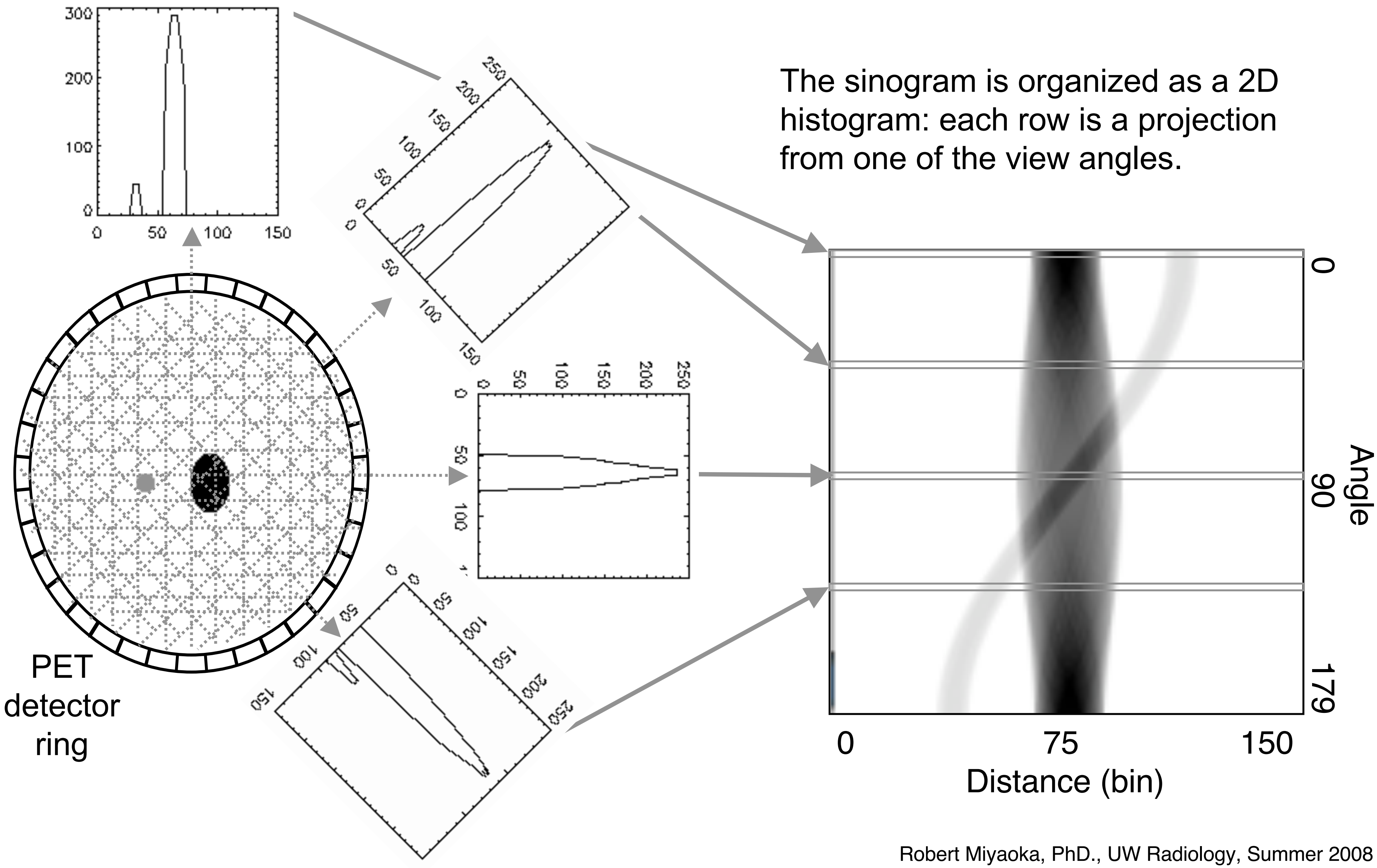
Image Reconstruction

- Takes raw sinogram from scanner and estimates underlying distribution (e.g. tracer concentration, tissue density)
- There are important user-specified control parameters that affect the noise/resolution trade-offs



Sinograms

Example of sinogram generation from one axial slice of an object volume



The sinogram is organized as a 2D histogram: each row is a projection from one of the view angles.

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Image Reconstruction: Backprojection

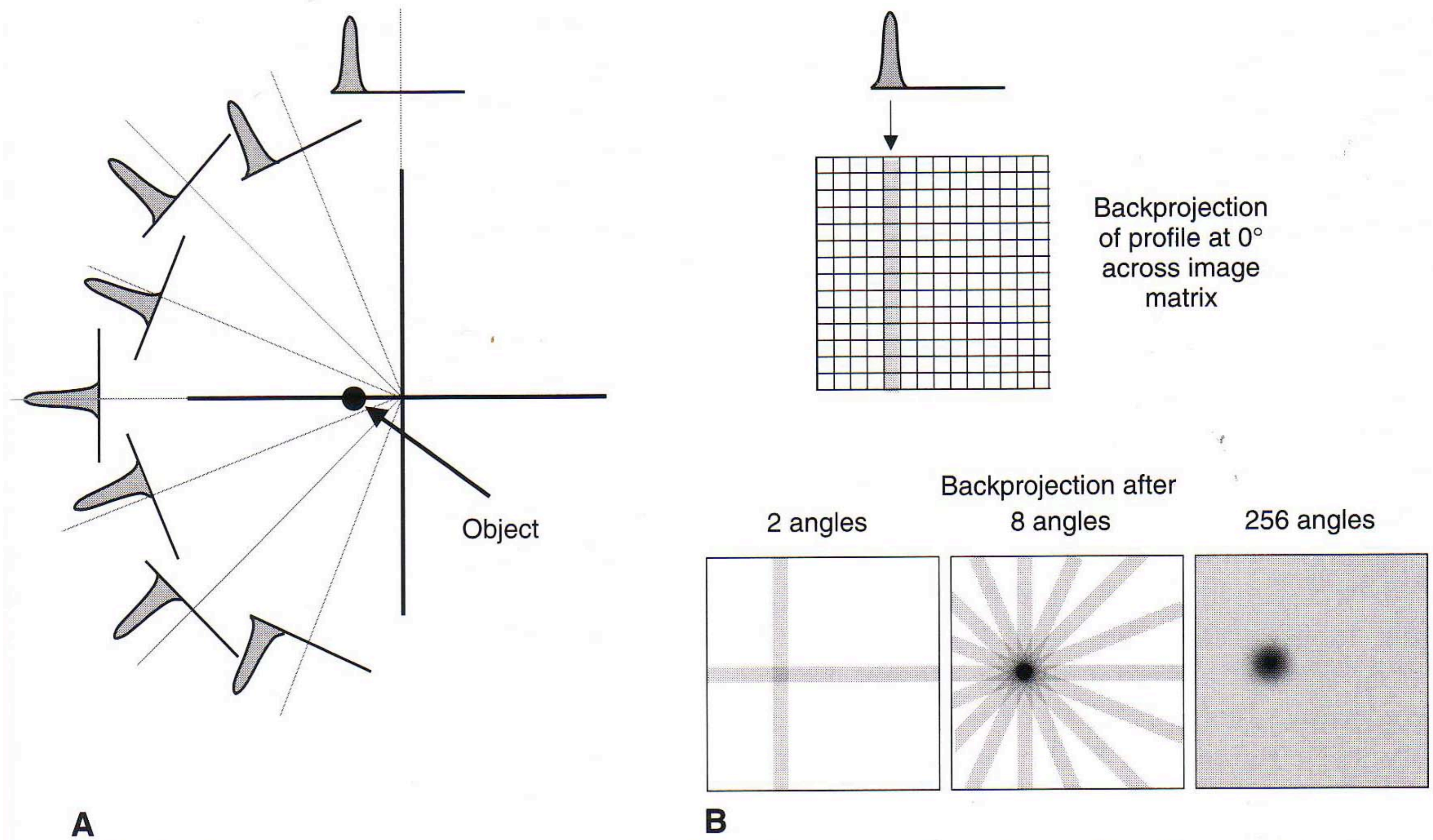
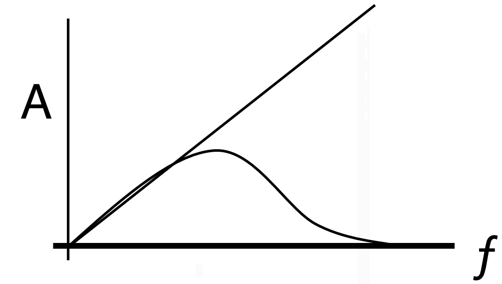


Figure 16–5. Illustration of the steps in simple backprojection. *A*, Projection profiles for a point source of radioactivity for different projection angles. *B*, Backprojection of one intensity profile across the image at the angle corresponding to the profile. This is repeated for all projection profiles to build up the backprojected image.

From: Physics in Nuclear Medicine (Cherry, Sorenson and Phelps)

Image Reconstruction: Filtered backprojection



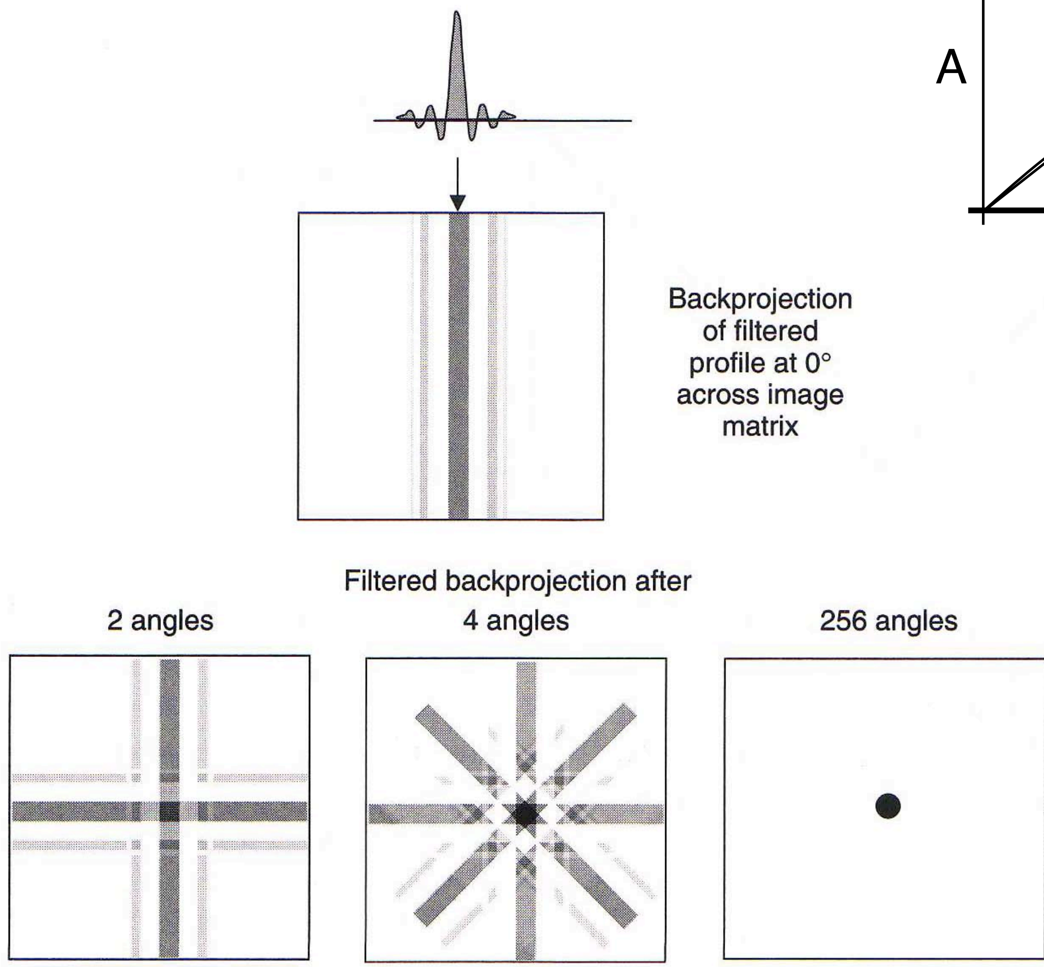
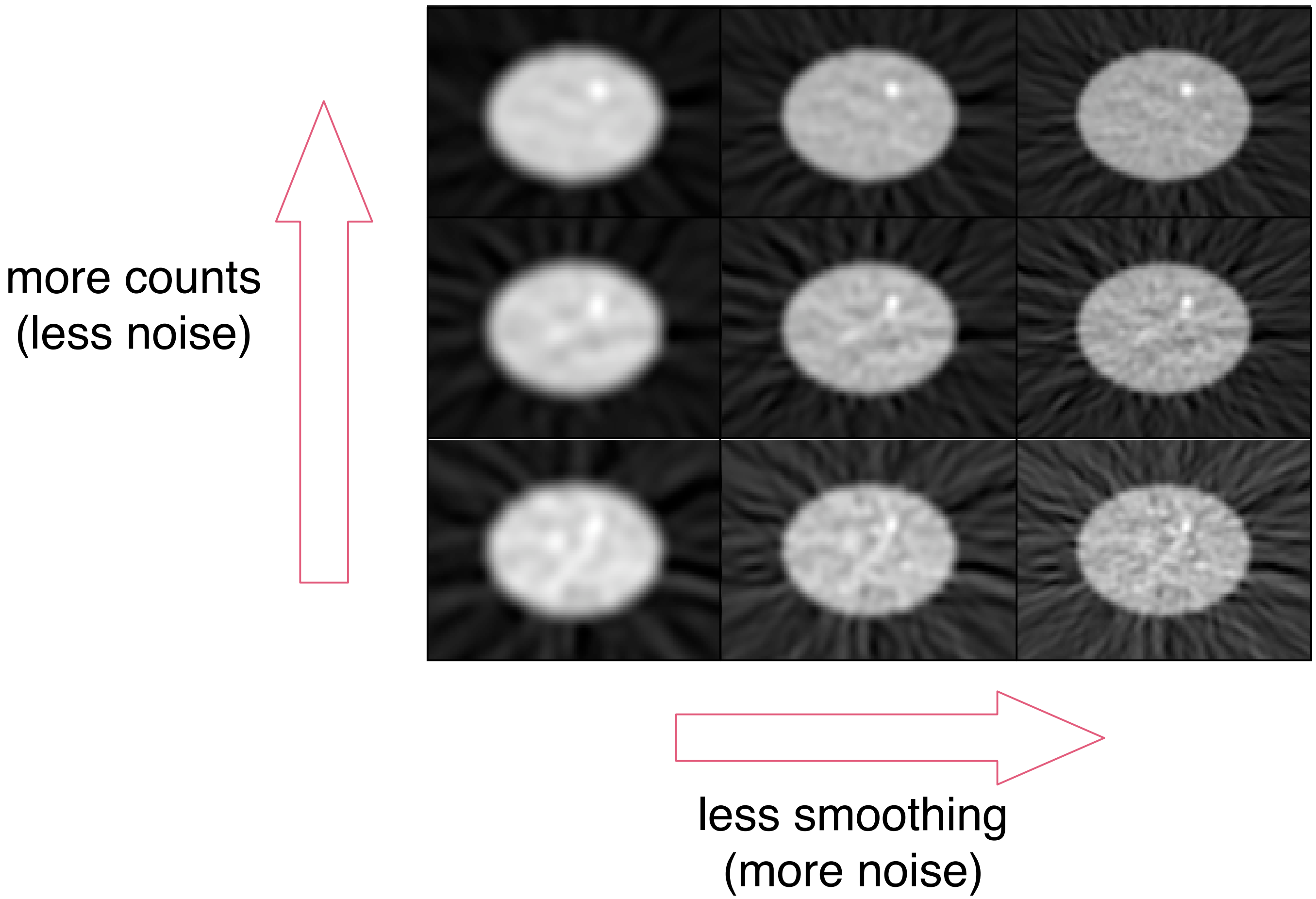


Figure 16–9. Illustration of the steps in filtered backprojection. The one-dimensional Fourier transforms of projection profiles recorded at different projection angles are multiplied by the ramp filter. After taking the inverse Fourier transform of the filtered transforms, the filtered profiles are backprojected across the image, as in simple backprojection.

From: Physics in Nuclear Medicine (Cherry, Sorenson and Phelps)

Effect of Smoothing vs. Noise with FBP

- Human abdomen simulation with 2cm diam. lesion 2:1 contrast



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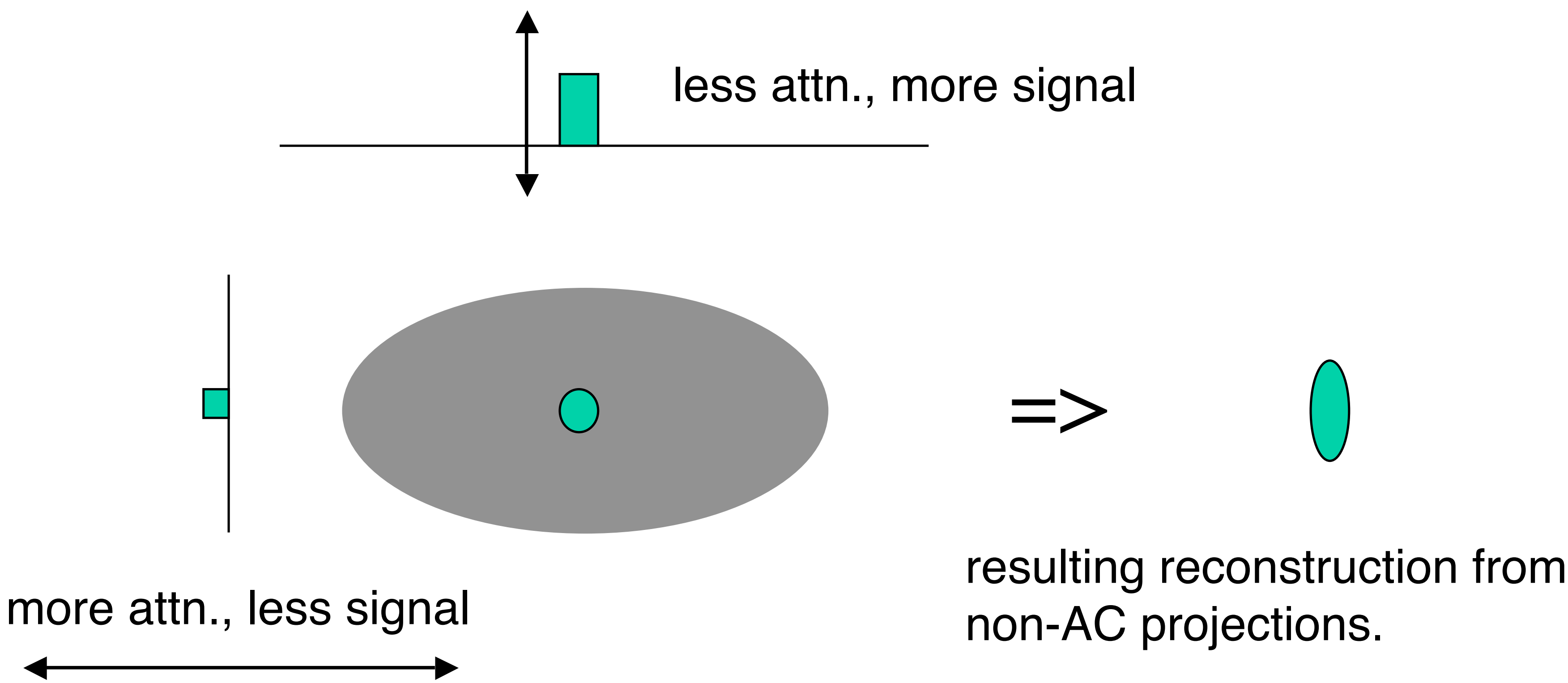
Attenuation Correction Errors

What happens if we do not apply attenuation correction?

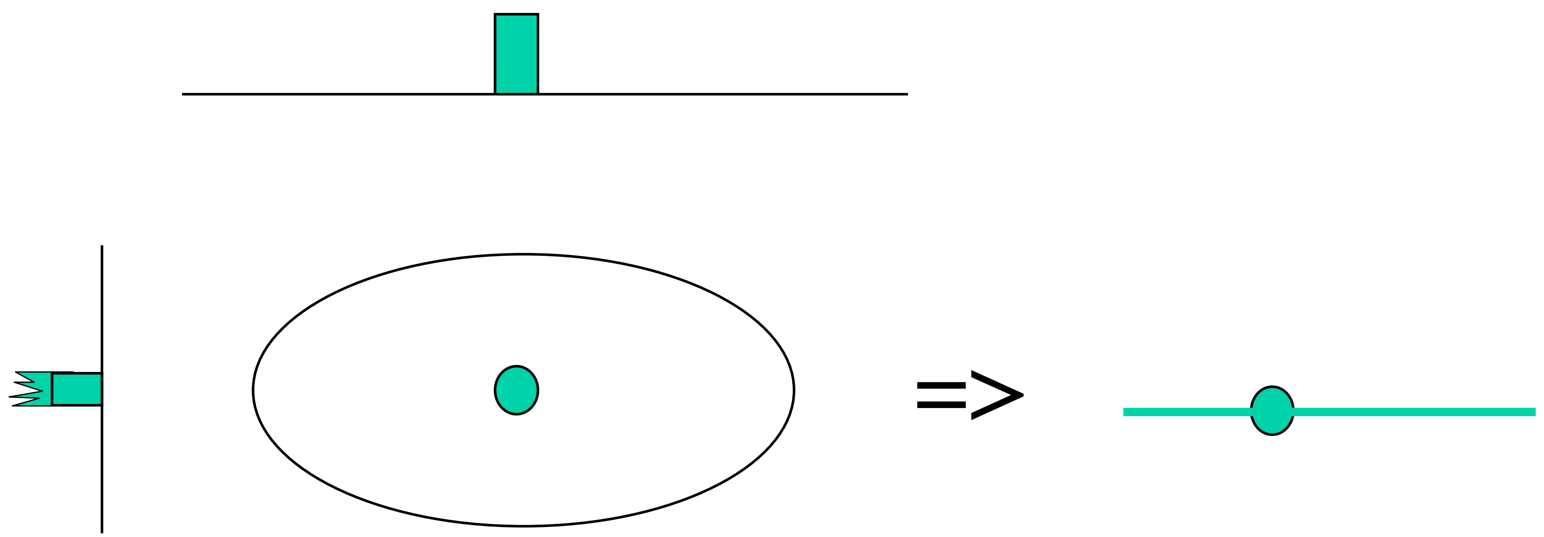
<u>Non quantitative values and distortions.</u>



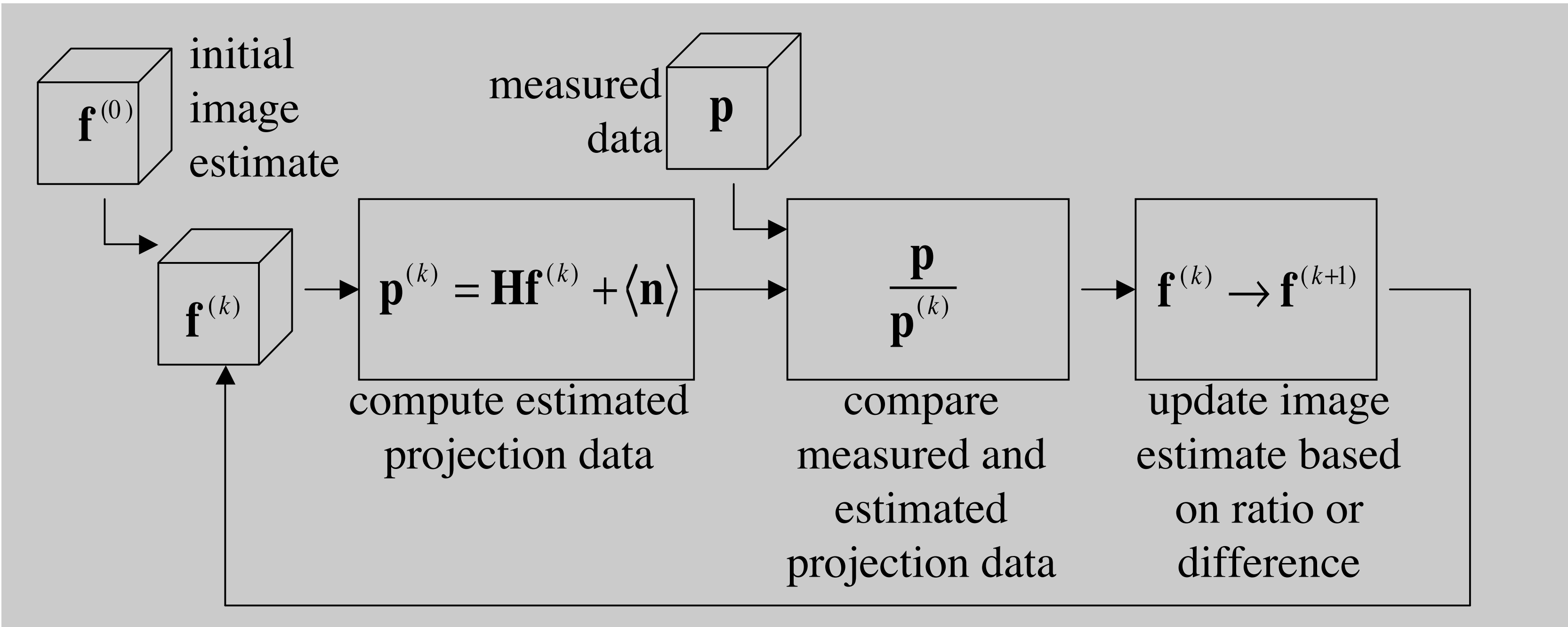
Attenuation Correction Errors

What happens if we do apply attenuation correction?

Streaks due to noise amplification for the low count projections, but get correct count densities.



A generic iterative procedure



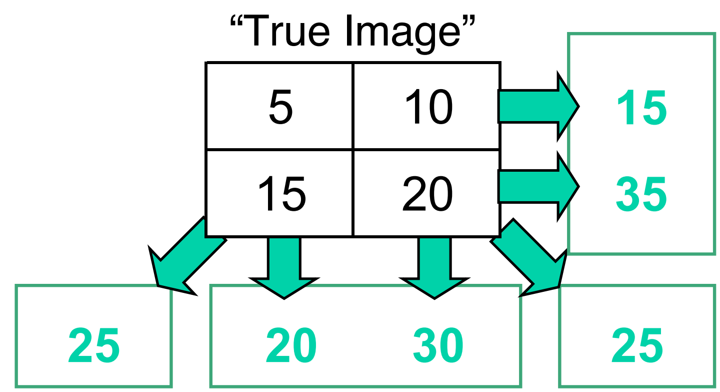
- There are many ways to:
  - model the system (and the noise)
  - compare measured and estimated projection data
  - update the image estimate based on the differences between measured and estimated projection data
  - decide when to stop iterating

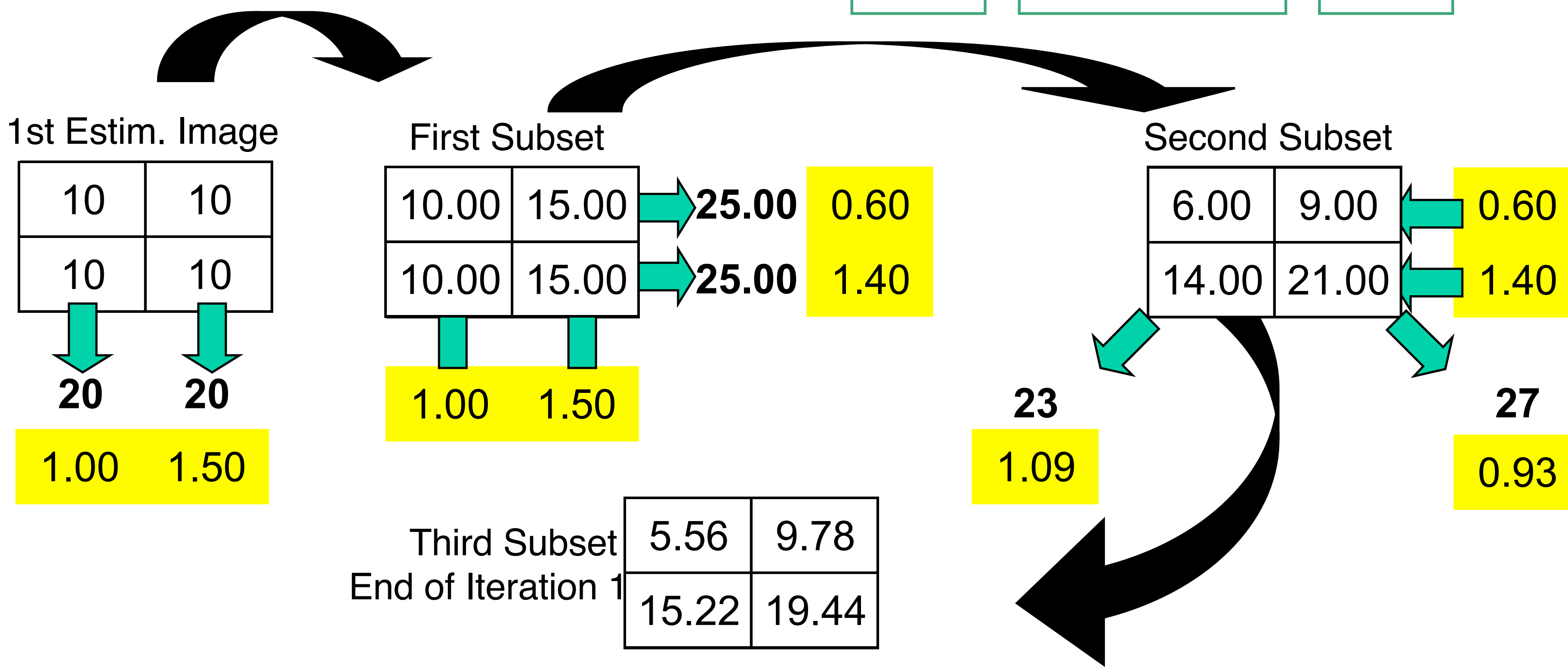
OS-EM Example

2 X 2 image: only 4 projection angles
(e.g. 0^o, 45^o, 90^o, 135^o)

Measured Data (ideal noiseless)

2 X 2 image: only 4 projection angles





Some test questions

D68. Positron cameras detect:

A. Positrons of the same energy in coincidence.
B. Positrons and electrons in coincidence.
C. Photons of different energies in coincidence.
D. Annihilation photons in coincidence.
E. Annihilation photons in anticoincidence.

D69. Spatial resolution of PET systems is determined by:

A. Detector size.
B. The ring diameter of the system.
C. The detector material.
D. Energy of the positron emitter in use.
E. All of the above.

D76. The spatial resolution of a SPECT image vs. a stationary image with the same camera is:

A. Much worse.
B. Slightly worse.
C. The same.
D. Slightly better.
E. Much better.

What about contrast resolution?

Same
Worse
Better

D77. The major limitation on the resolution of an FDG scan on a modern whole body PET scanner is:

A. Range of the positron.
B. Image matrix size.
C. The physical size of the individual detectors.
D. The non-collinearity between the annihilation photons.
E. Attenuation correction.

D78. A nuclear medicine resident discovers, just prior to injecting a Tc-99m bone scan agent, that the patient had a PET scan 3 hours ago at 9 a.m. in another hospital. When should the resident recommend that the bone scan be performed?

A. Straight away. There is no interference between the Tc-99m and F-18, since they can be distinguished by energy discrimination.
B. Wait until 3 p.m. allowing a 6-hour interval between tests (>3 half lives of F-18).
C. Wait until the next day to ensure complete decay of the F-18.
D. Postpone for one week, to ensure any residual long lived F- 18 daughters have decayed.

D77. Some dedicated PET scanners can perform both 2-D and 3-D scans. The difference is:

A. 2-D scans acquire transaxiai images and cannot display coronal or sagittal images.
B. 3-D scans acquire the data directly in coronal or sagittal planes.
C. 2-D scans acquire the data one slice at a time, whereas 3D scans acquire all slices simultaneously.
D. Only 3-D scans can be corrected for attenuation.
E. 2-D scans have septa in front of the detectors to reduce events from scattered photons.

D79. The assigned values in each pixel in the reconstructed image of SPECT represent:

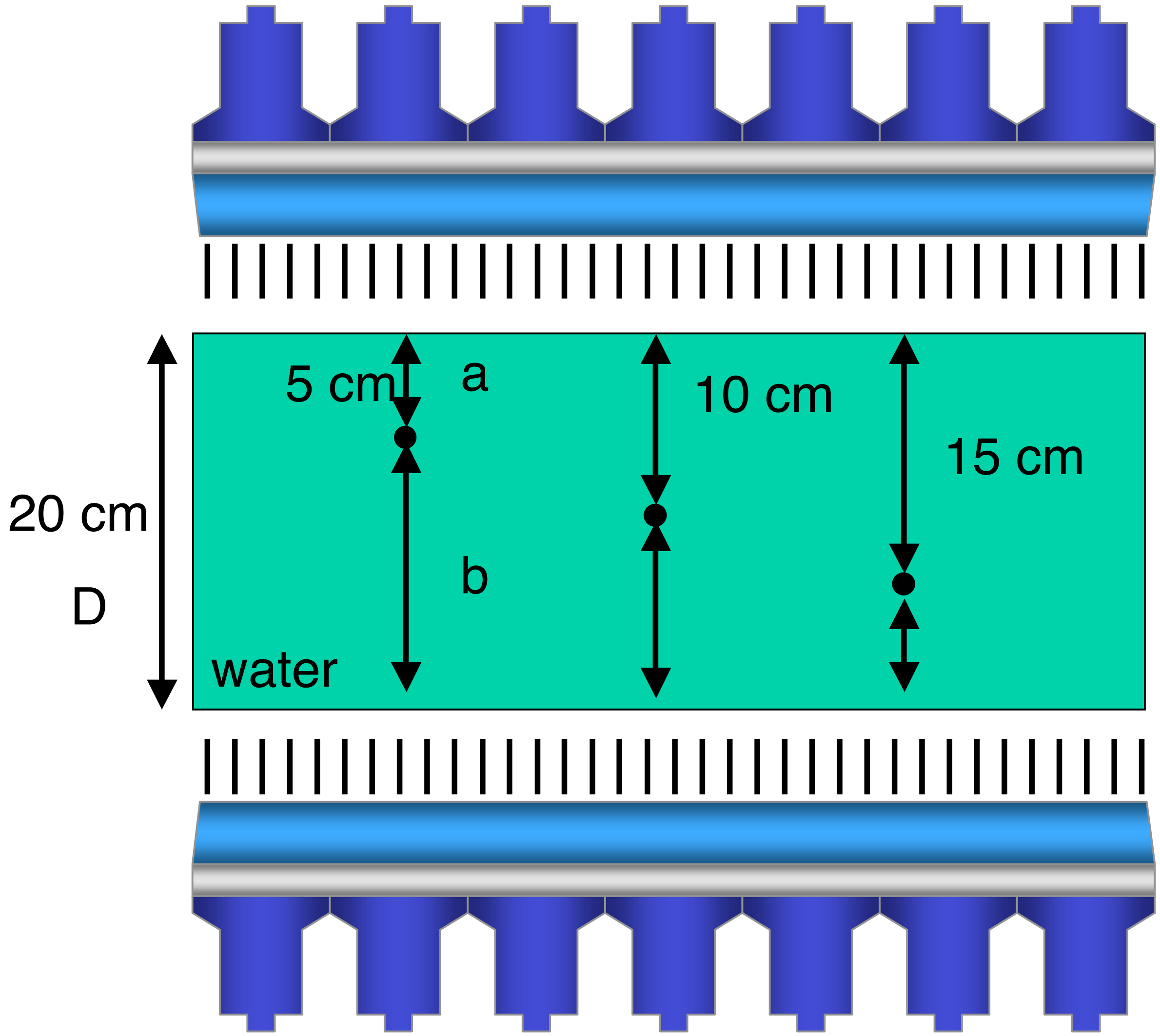
A. Densities.
B. Absorption factors.
C. Attenuation factors.
D. Radioisotope concentrations.

D85. All of the following are true statements about PET scanning, ***except:***

A. Radioisotopes are cyclotron produced.
B. Positrons are not detected directly.
C. Coincident detection at 180° is required.
D. Images are generally axial tomograms.
E. The detector photopeak is centered at 1.02 MeV.

Extra stuff

SPECT: Conjugate counting



Arithmetic Mean:

$I_A = (I_1+I_2)/2$

Geometric Mean:

$I_G = (I_1 x I_2)^{1/2}$

reduces depth dependent effects

SPECT: Arithmetic vs. Geometric Mean

$I_{det1} = I_0(e^{-\mu a})$ $\qquad$ $I_{det2} = I_0(e^{-\mu b})$

Arithmetic Mean:

$$I_A = (I_{det1}+I_{det2})/2 = (I_0(e^{-\mu a}) + I_0(e^{-\mu b}))/2$$

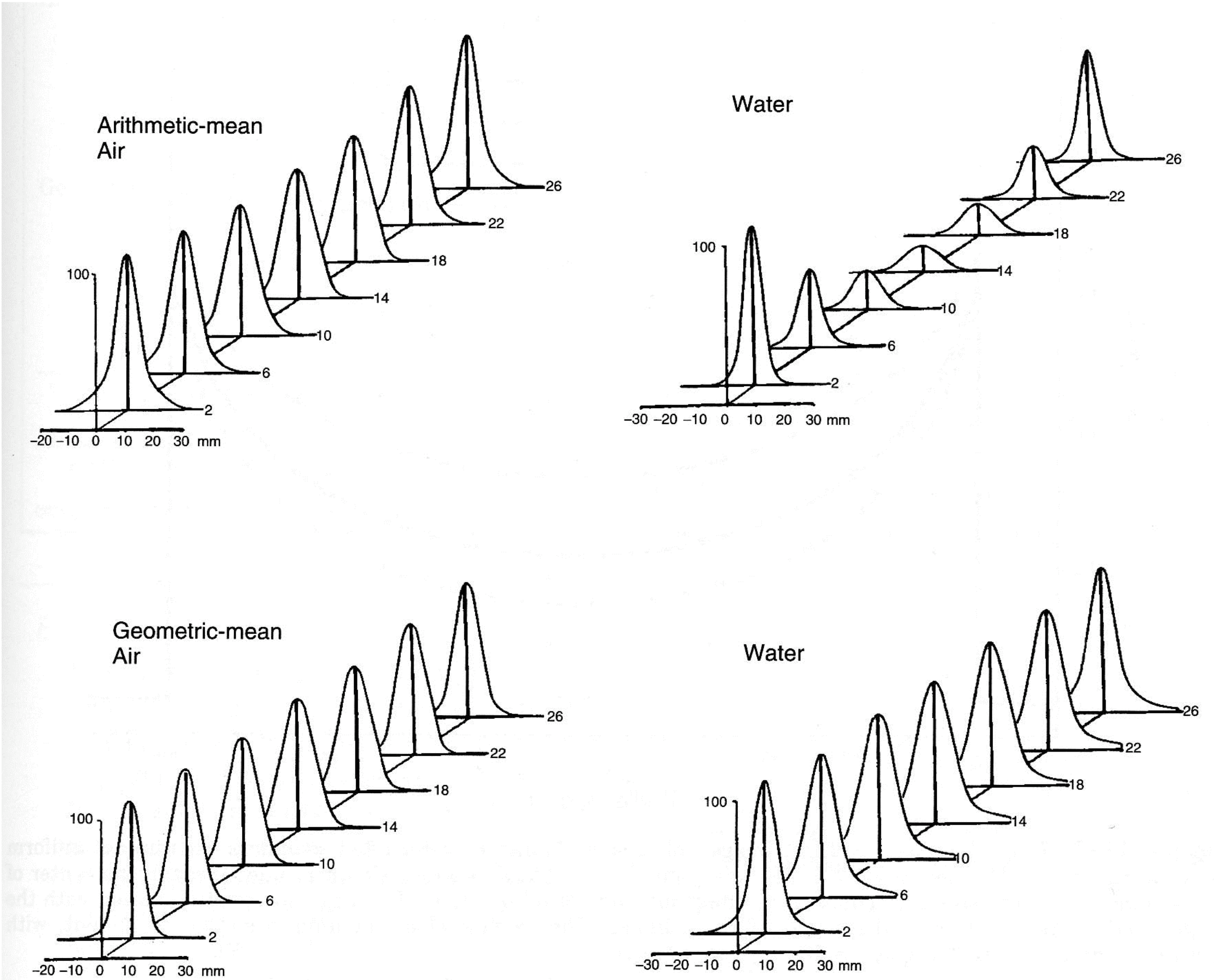
Geometric Mean:

$$I_G = (I_{det1} x I_{det2})^{1/2} = ((I_0 x I_0)e^{-\mu x(a+b)})^{1/2}$$

$$= (I_0 x I_0)^{1/2} e^{-\mu x(a+b)/2}$$

$$= I_0 \, e^{-\mu D/2}$$

Arithmetic vs. Geometric Mean

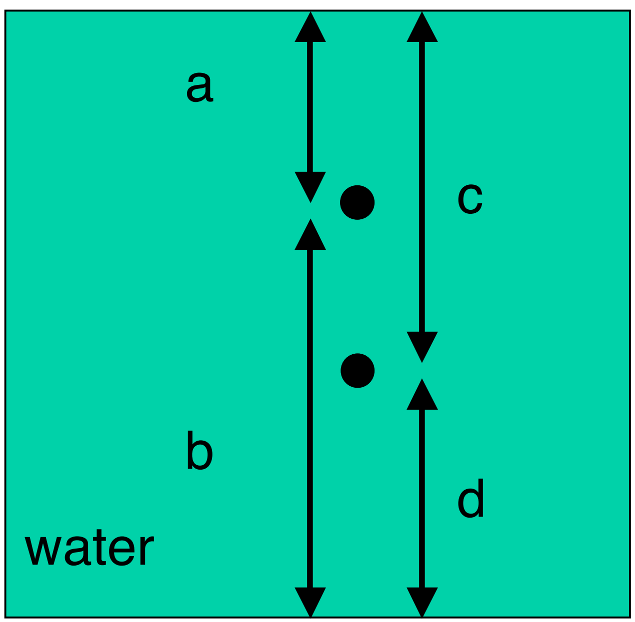


From: Physics in Nuclear Medicine (Cherry, Sorenson and Phelps)

Geometric Mean

Unfortunately, simple formula to compensate for depth dependence breaks down for more complicated objects.

detector



detector

$$I_1 = I_{01}e^{-\mu a} + I_{02}e^{-\mu c}$$

$$I_2 = I_{01}e^{-\mu b} + I_{02}e^{-\mu d}$$

$$I_G = [(I_{01}^2 + I_{02}^2)e^{-\mu D} + I_{01}I_{02}e^{-\mu(a+d)} + I_{01}I_{02}e^{-\mu(c+b)}]^{1/2}$$

Chang Attenuation Correction

* Can be used with FBP
* Post reconstruction correction
* Often assume uniform attenuation

$$ACF(x,y) = \frac{1}{\frac{1}{N}\sum_{i=1}^{N} e^{-\mu d_i}}$$

Determine average ACF over all projection angles for each pixel

$$f(x,y) = f'(x,y) \times ACF(x,y)$$

Multiply initial image by ACF's to get initial guess of real distribution

$$p_{\text{error}}(r,\phi) = p(r,\phi) - p_{\text{fp}}(r,\phi)$$

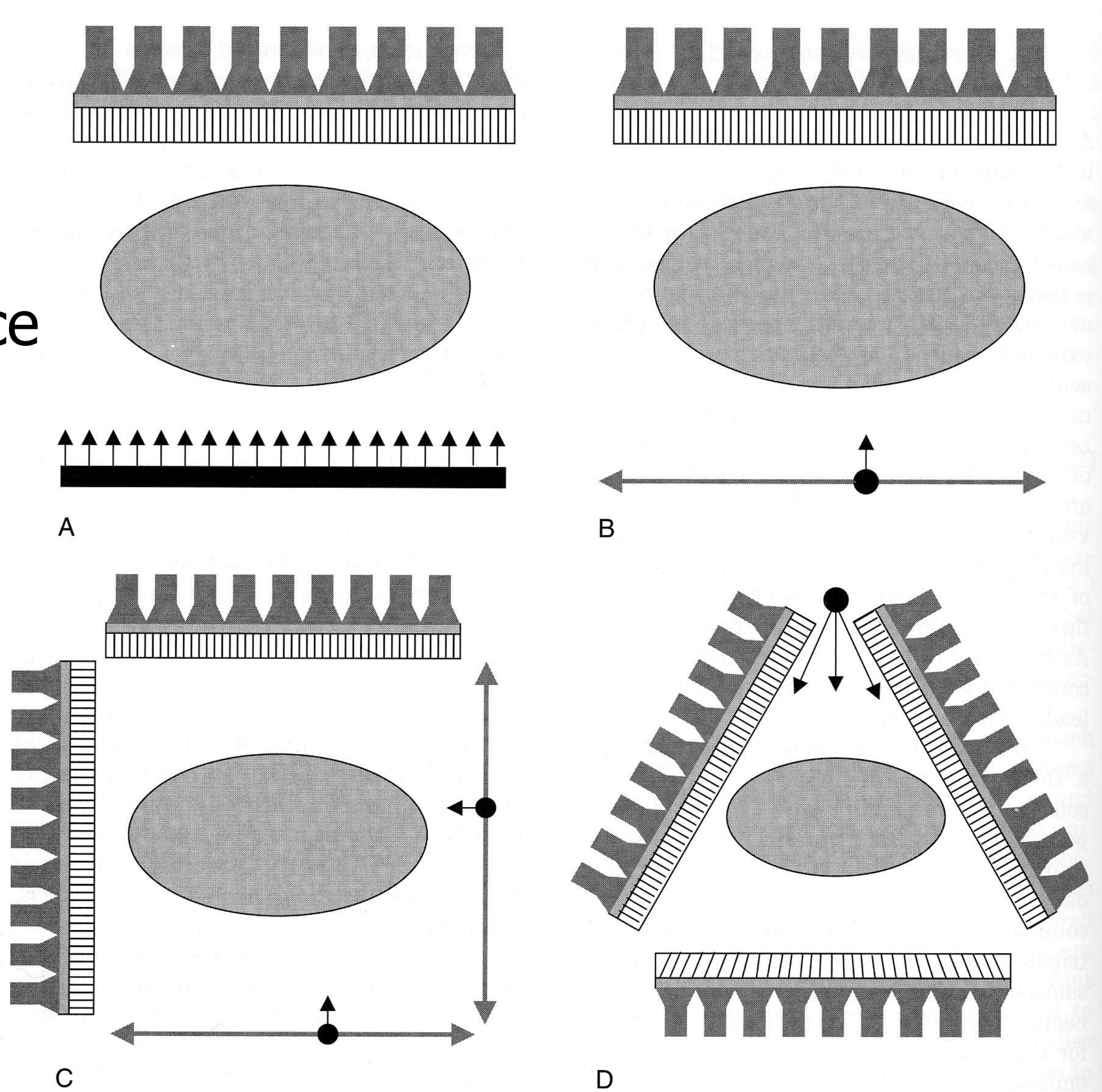
Forward project *f(x,y)* including atten and subtract from acq data

$$f(x,y) = f'(x,y) \times ACF(x,y) + f_{\text{error}}(x,y) \times ACF(x,y)$$

Reconstruct error image then subtract from initial image

From: Physics in Nuclear Medicine (Cherry, Sorenson and Phelps)

SPECT: Transmission Scans



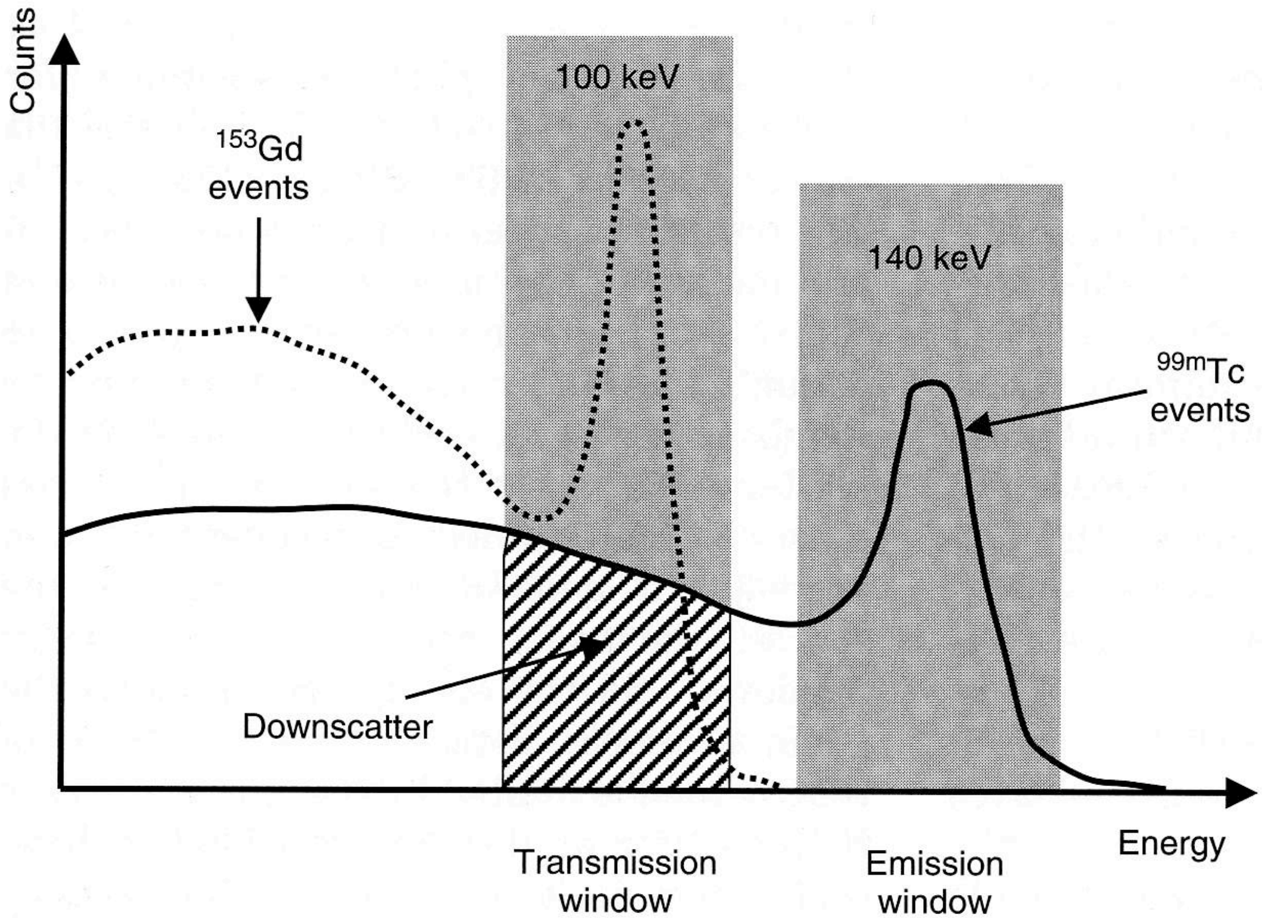
Collimated flood source

Scanning line source

Scanning line source

Fixed line source with converging collimator

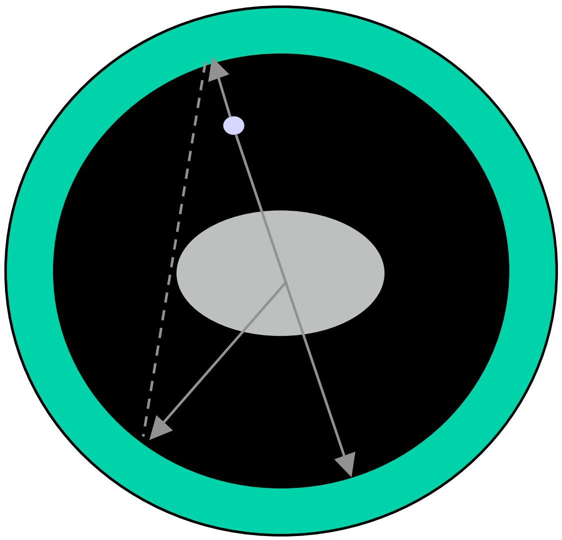
SPECT: Transmission Data



Depending upon istotope used may have to account for down scatter
Especially for simultaneous emission-transmission imaging

Gated coincidence transmission

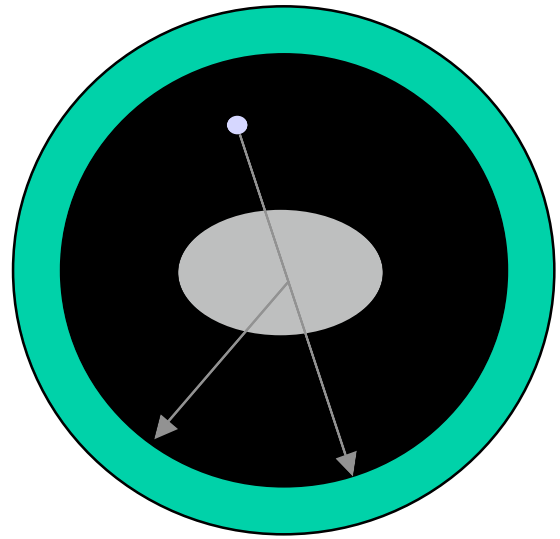
Coincidence + gating rejects most scatter



Problem - count rate limited by detector closest to rod source

Singles transmission

Count rate limited by the far detector



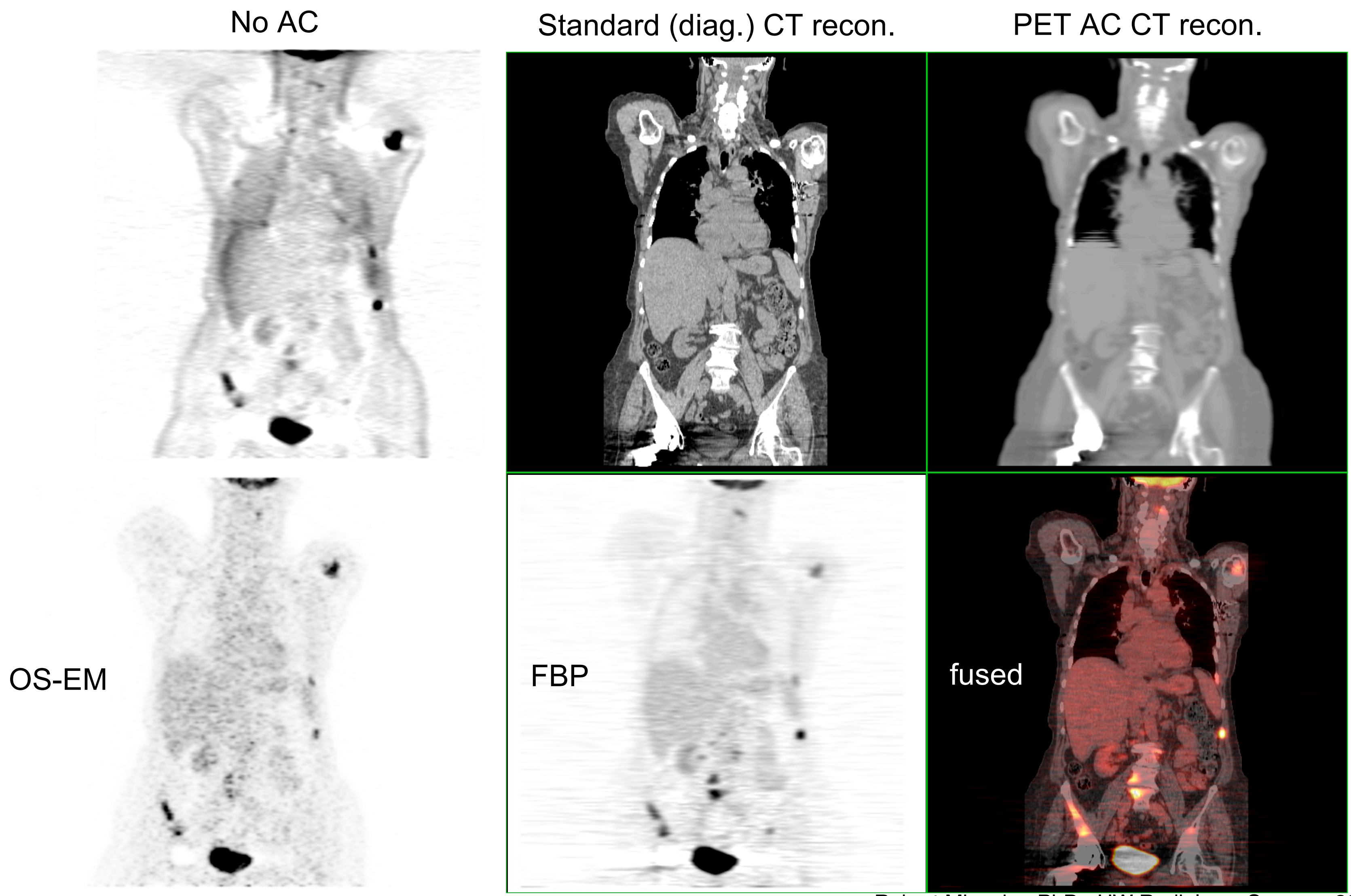
Problems - scatter; hardware modifications to tomograph

Solution - use image segmentation on attenuation data sets

From: Lewellen fall 2005

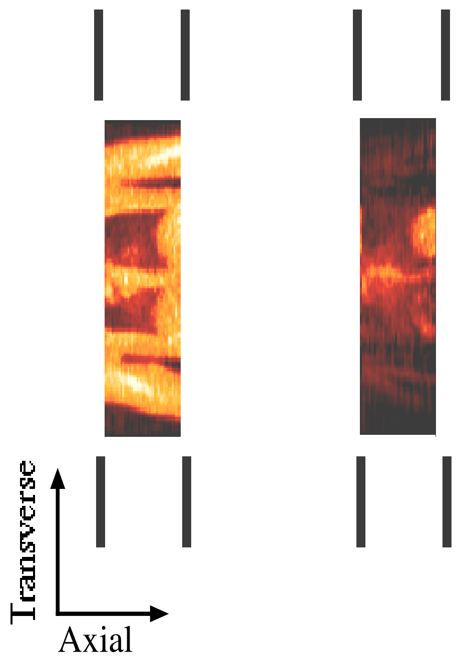
PET Attenuation Correction (CTAC)

PET AC is now performed with a CT image (CT can also be used for SPECT)



Robert Miyaoka, PhD., UW Radiology, Summer 2008

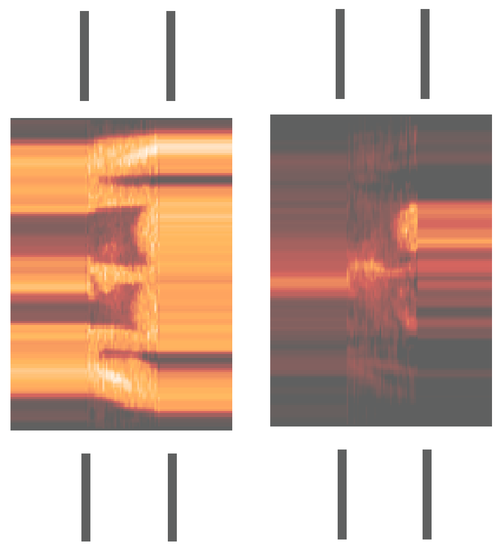
Three ways to estimate out of field activity



Only use FOV data

Pro: simple

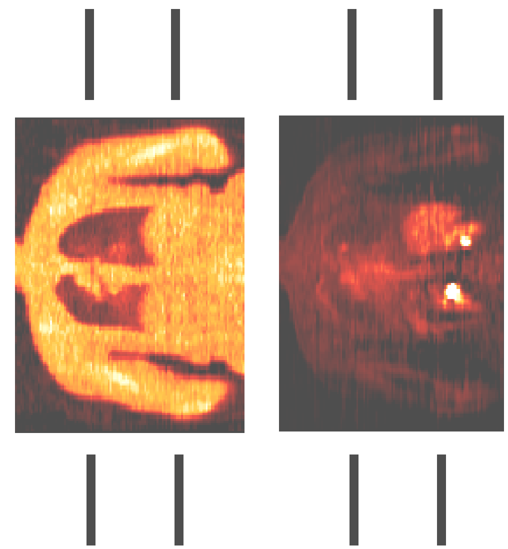
Con: Subject to bias



Extend end planes

Pro: easy to implement

Con: may not work if activity changes rapidly



Include over scan data

Pro: Samples external FOV data

Con: More acquisitions