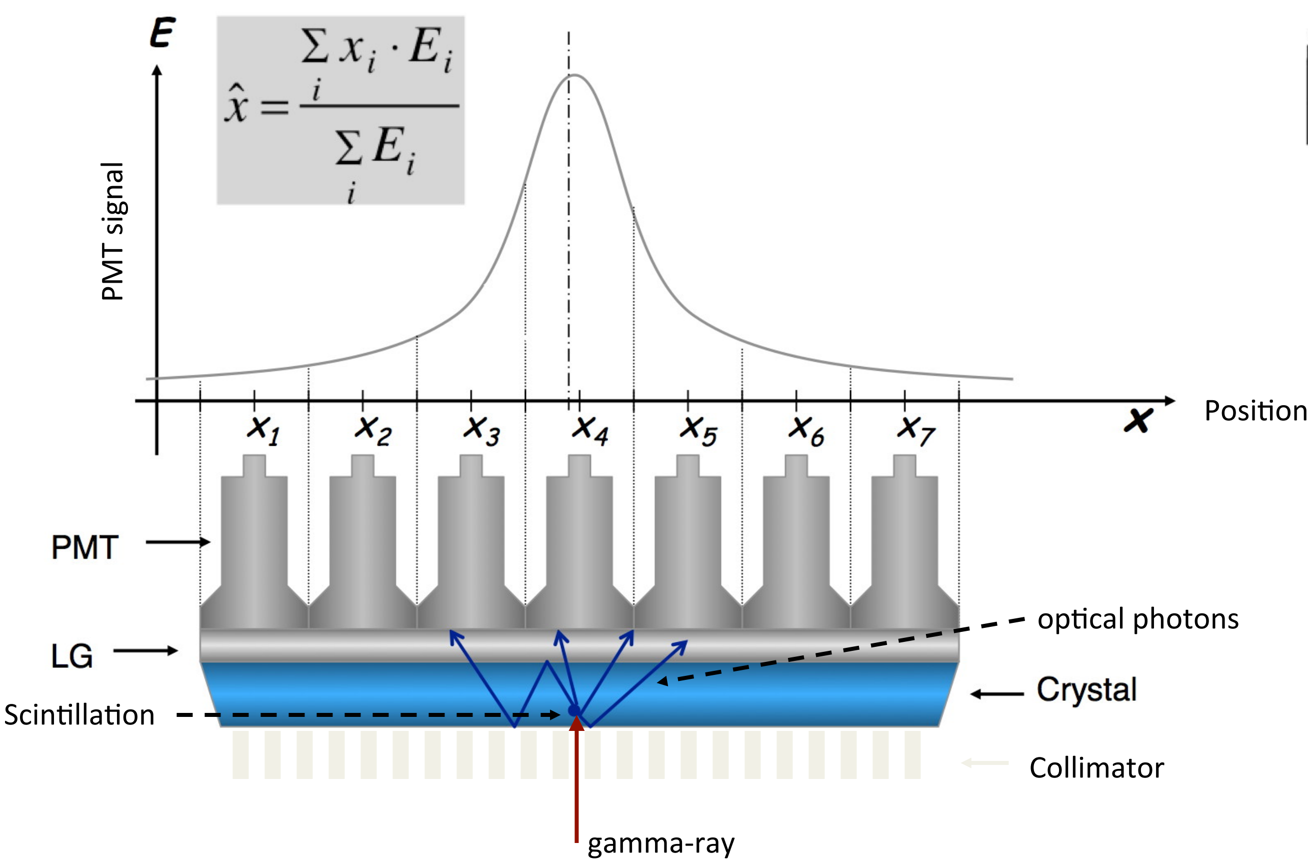
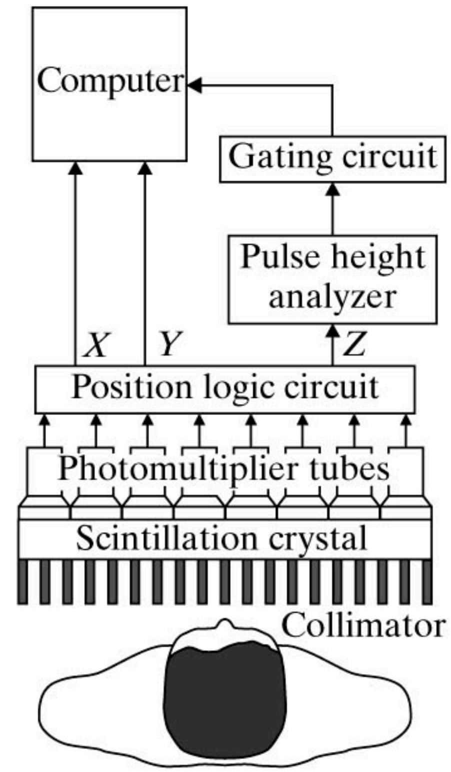
Nuclear Medicine Imaging Systems

- Position estimation
  - based on 2-D centroid calculation
  - called Anger logic or Anger camera





Analogies with X-ray Imaging

X-rays

- photon formation
  - bremsstrahlung
  - characteristic x-rays
- attenuation by tissues
  - photoelectric absorption
  - Compton scatter
- detection systems
  - scintillators
- 2-D projection imaging
  - radiographs
- tomographic imaging
  - CT
  - collect projections from around the patient

Nuclear medicine

- photon formation
  - positron emission
  - isomeric transition
- attenuation by tissues
  - photoelectric absorption
  - Compton scatter
- detection systems
  - scintillators
- 2-D projection imaging
  - scintigraphs
- tomographic imaging
  - SPECT and PET
  - collect projections from around the patient

Gamma Camera Imaging (SPECT)

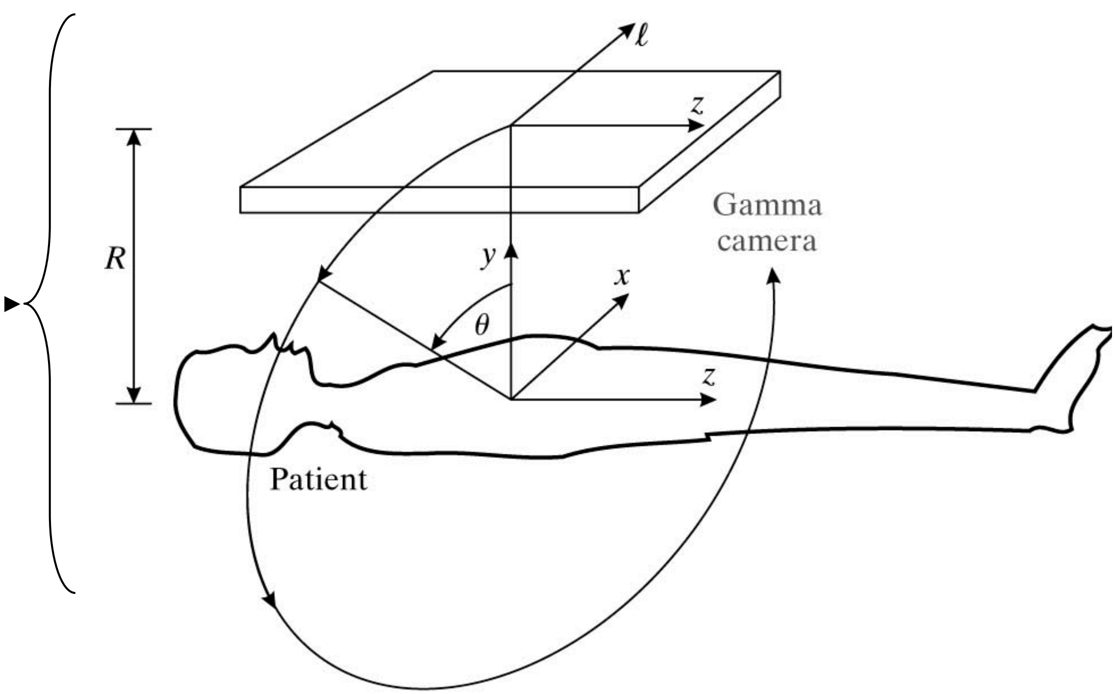
Siemens Healthcare, Symbia

Philips Medical, Brightview XCT

GE Healthcare, MG

Single Photon Emission Computed Tomography

- SPECT is to scintigraphy as CT is to radiography
  - We collect projection views all the way around the patient
  - We use tomographic principles (e.g. FBP) to reconstruct an image
  - One difference is that we are using emission rather than transmission
  - Another difference is that attenuation is a headache rather than what we want (as for CT)
  - Scattered photons, however, remain a nuisance



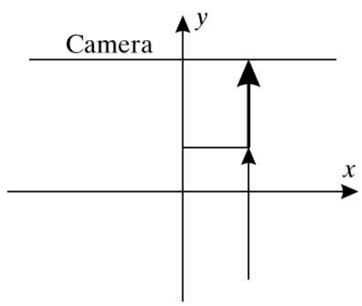
SPECT Imaging Equation

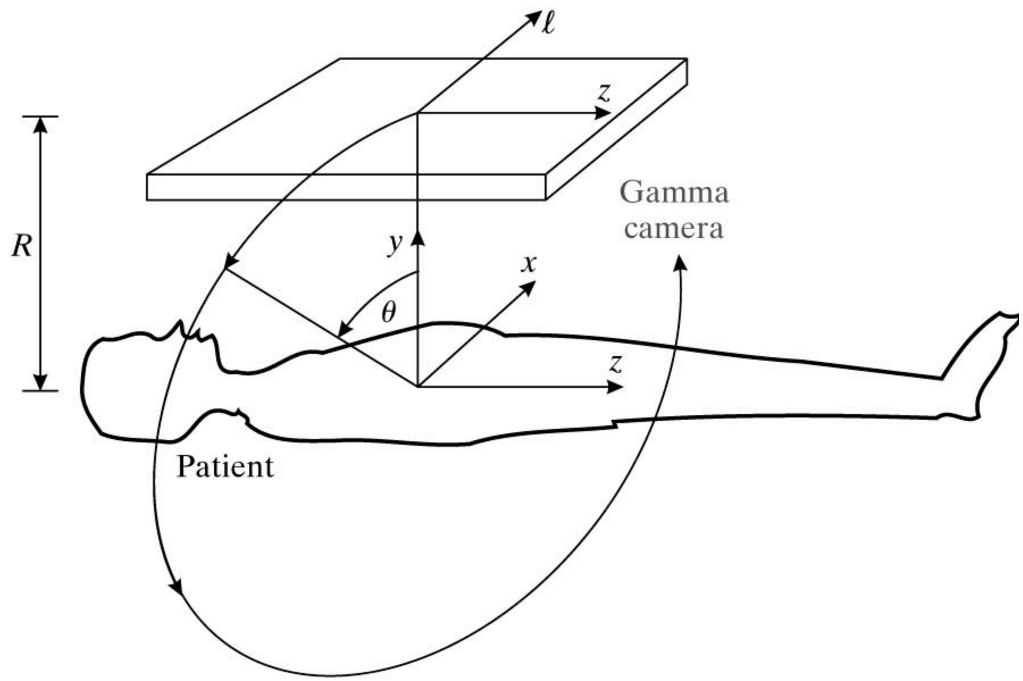
- Using parallel-hole collimators, and ignoring depth-dependent collimator burring etc., we have

$$\phi(z,l) = \int_{-\infty}^{R} \frac{A(x,y,z)}{4\pi(y-R)^2} \exp\left\{-\int_{y}^{R} \mu(x,y',z,E)\,dy'\right\} dy$$

- Our inverse problem is: given $\phi(z,l)$ for all angles θ, what is $A(x,y,z)$?
  - Like poly-energetic CT problem this is hard
  - Complicated by attenuation
  - Solution discovered in 2000

integration geometry

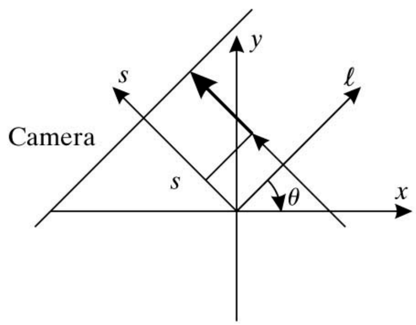




SPECT Image Reconstruction

- For the approximate solution used in practice, we make several assumptions
  - ignore collimator effects
  - parameterize x and y as functions of distance along line

integration geometry



$$x(s) = l\cos\theta - s\sin\theta$$

$$y(s) = l\sin\theta + s\cos\theta$$

- This changes the imaging equation to:

$$\phi(l,\theta) = \int_{-\infty}^{R} \frac{A(x(s),y(s))}{4\pi(s-R)^2} \exp\left\{-\int_{s}^{R} \mu(x(s'),y(s');E)\,ds'\right\} ds$$

SPECT Image Reconstruction

- Additional assumptions
  - ignore inverse-square dependence of fluence
  - assume we can correct for attenuation effects later
- This changes the imaging equation to

$$\phi(l,\theta) = \int_{-\infty}^{\infty} A(x(s),y(s))\,ds$$

$$= \int_{-\infty}^{\infty}\int_{-\infty}^{\infty} A(x,y)\delta(x\cos\theta + y\sin\theta - l)\,dx\,dy$$

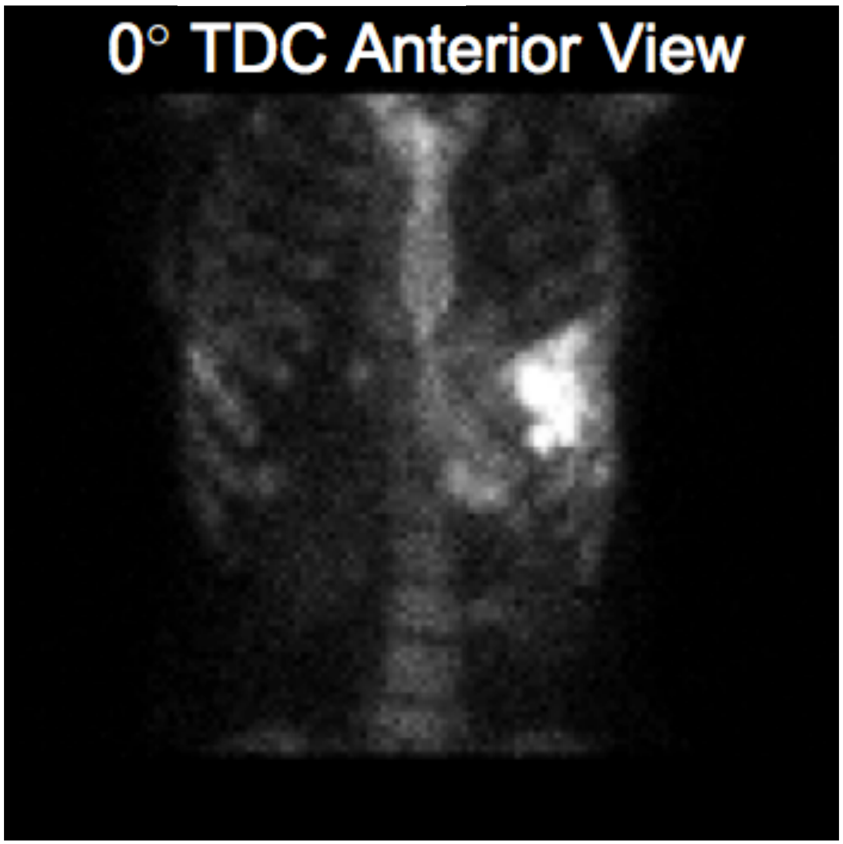
which is the 2-D x-ray or Radon transform that we can solve with FBP

$$A(x,y) \simeq \int_0^{\pi}\left[\int_{-\infty}^{\infty}|\rho|\Phi(\rho,\theta)e^{j2\pi\rho l}\,d\rho\right]d\theta$$

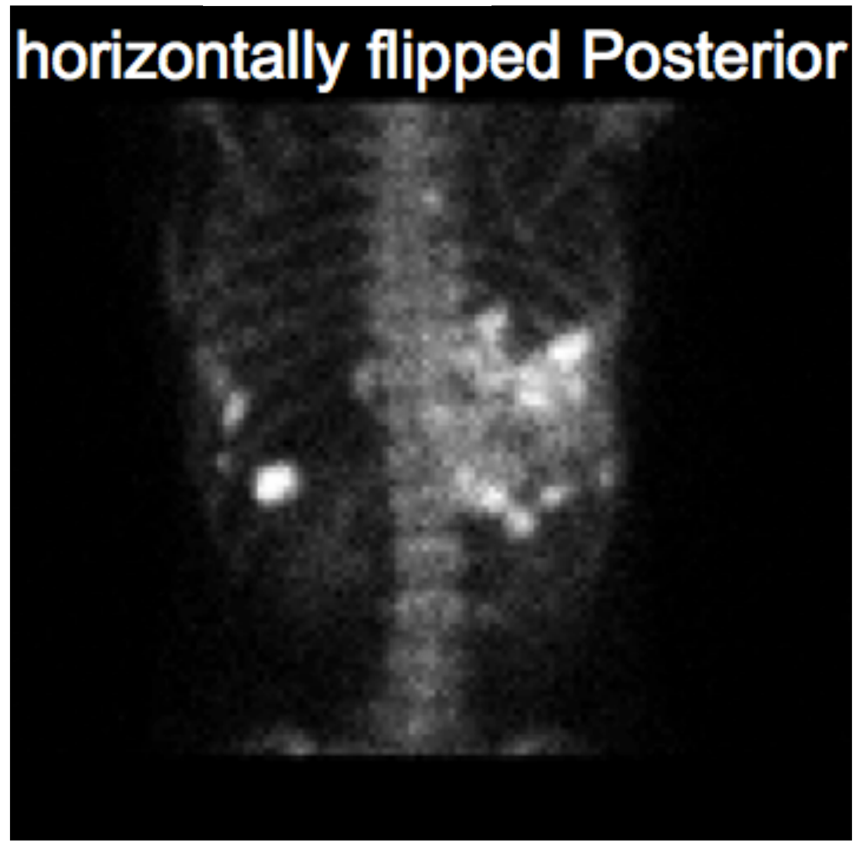
where $\Phi(\rho,\theta) = \mathrm{F}_{1\mathrm{D}}\{\phi(l,\theta)\}$

Angular asymmetry

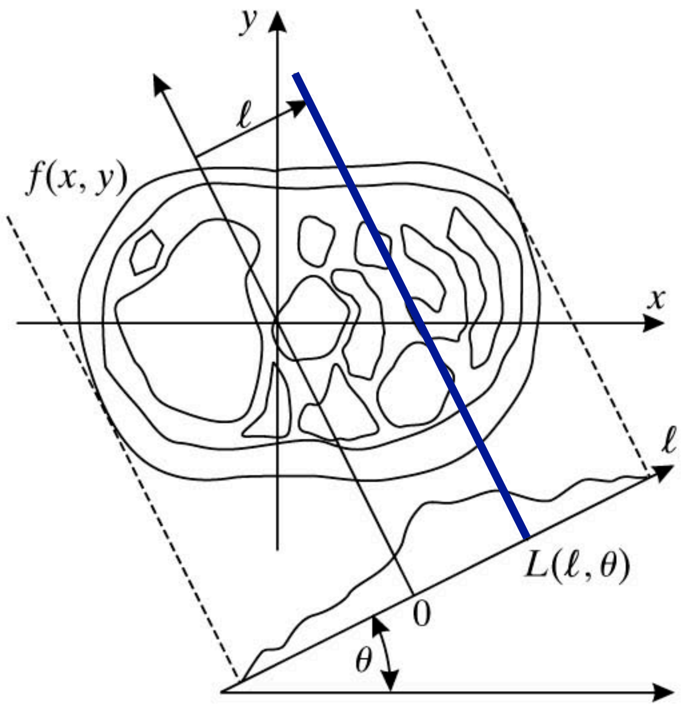
$\phi(l,\theta)$



$\phi(l,\theta +p)$



Comparison of Imaging Equations



x-ray transform integral along line

$$L(l,\theta) = \left\{(x,y) \middle| x\cos\theta + y\sin\theta = l\right\}$$

With rotated coordinates (l,s)

$$x(s) = l\cos\theta - s\sin\theta$$

$$y(s) = l\sin\theta + s\cos\theta$$

CT $$\phi(l,\theta) = \int_0^{E_{\max}} S_0(E)E\exp\left\{-\int_{-R}^{R}\mu(x(s'),y(s');E)\,ds'\right\}dE$$

SPECT $$\phi(l,\theta) = \int_{-\infty}^{R}\frac{A(x(s),y(s))}{4\pi(s-R)^2}\exp\left\{-\int_{s}^{R}\mu(x(s'),y(s');E)\,ds'\right\}ds$$

PET $$\phi(l,\theta) = K\int_{-R}^{R}A(x(s),y(s))\,ds\cdot\exp\left\{-\int_{-R}^{R}\mu(x(s),y(s);E)\,ds\right\}$$

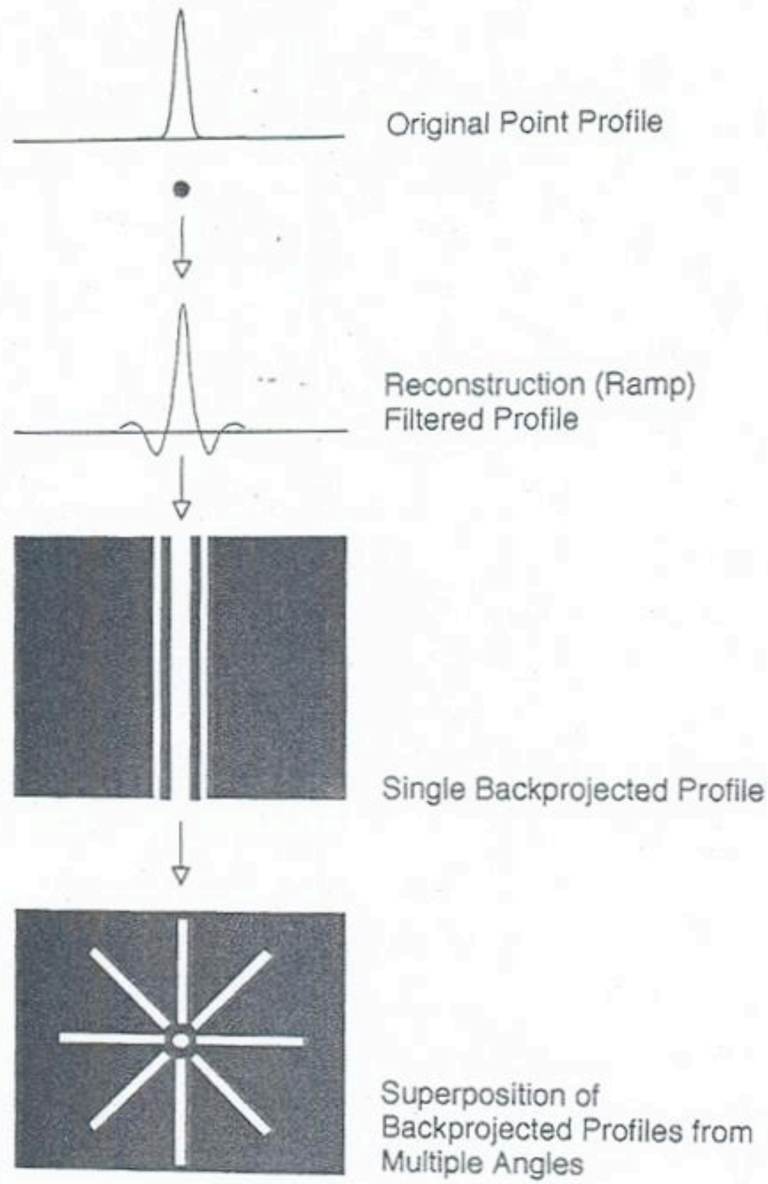
Image reconstruction

Image reconstruction

Point source reconstrction by
Filtered back projection (FBP)

Filtering the projection data with a ramp filter creates the negative values in back-projected profile necessary for cancelling the contributions from other angles in the region about the reconstructed point.

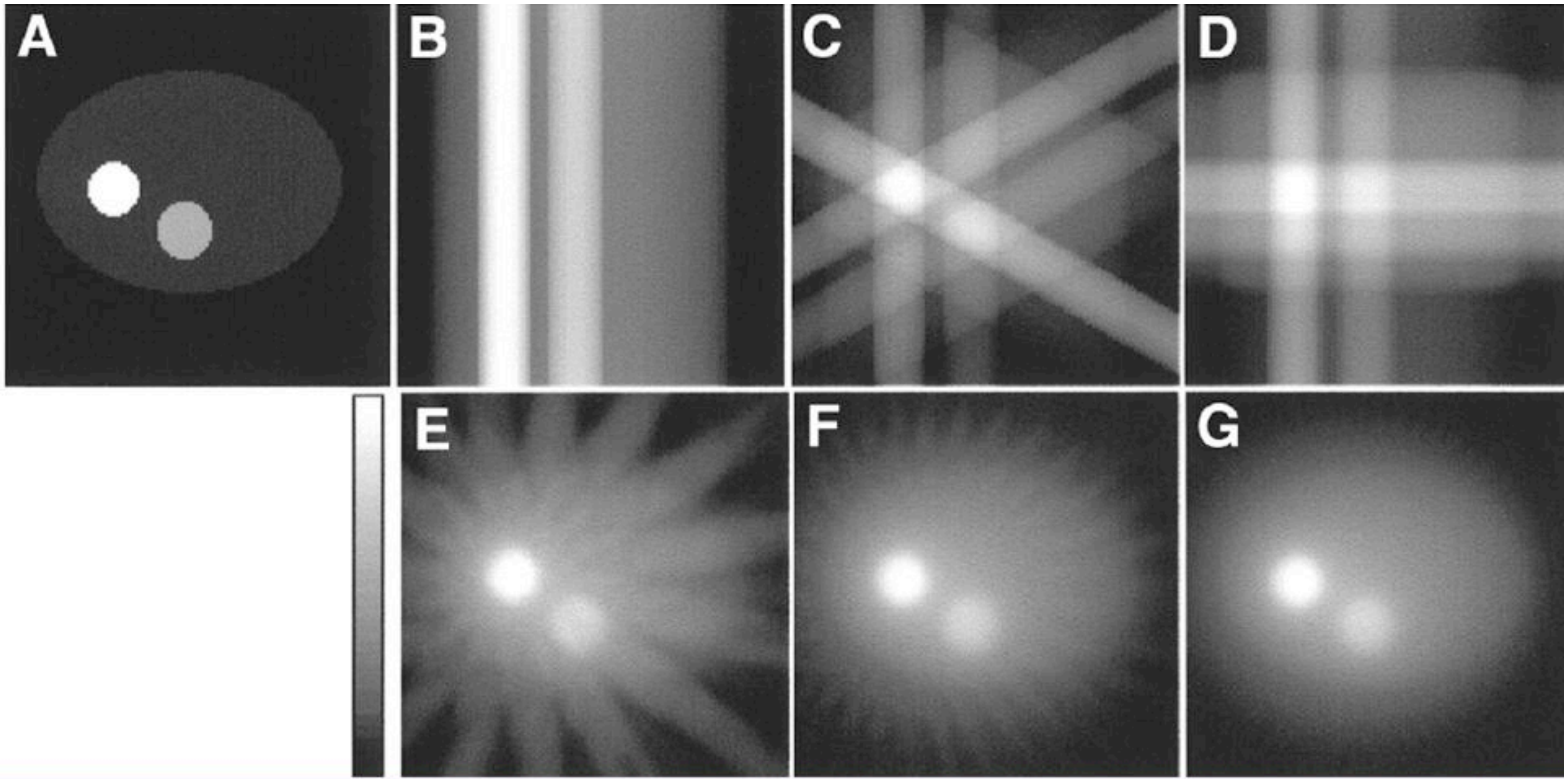
The back projections from different angles are added together to form the reconstructed image



Cullom SJ, In: Cardiac SPECT Imaging, 2001

Image reconstruction

Star (or streak) artifact of FBP reconstruction



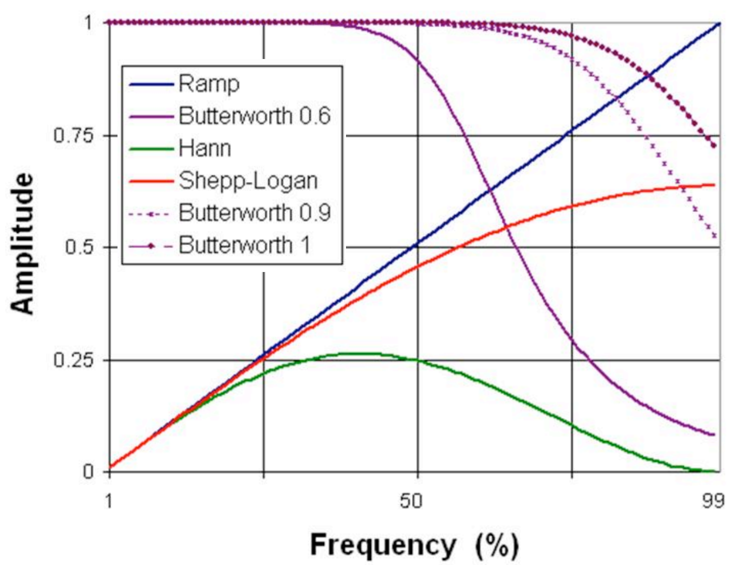
- A = object
- B through G = {1, 3, 4, 16, 32, 64} number of projections
- Star artifact decreases with number of projections

Bruyant PP, *J Nucl Med* 2002; 43:1343–58

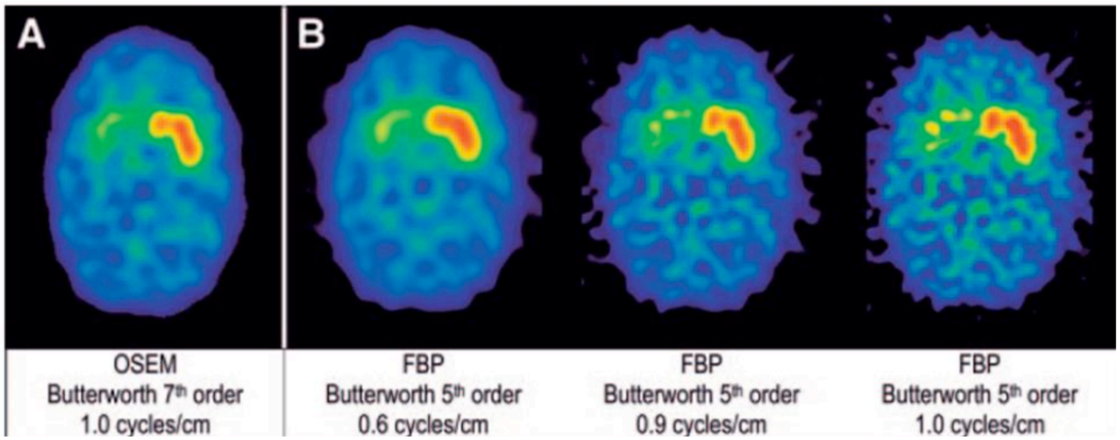
Reconstruction Filters
used to reduce image noise

- A ramp filter of the form “|r|” used to correct for “1/r” tomographic blurring artifact
- Often, the “ramp” filter is further modified by additional frequency filters, such as: the “Hann,” “Shepp-Logan” or “Butterworth” filters.
- In filtering, we see a tradoff of image noise vs. spatial resolution
- Higher cutoff frequencies maintain spatial resolution at the expense of more noise
- Filtered FBP is a very rapid technique for reconstructing SPECT data
- Iterative reconstruction techniques, such as ordered subset will be covered in later lecture

Modified Frequency Filters

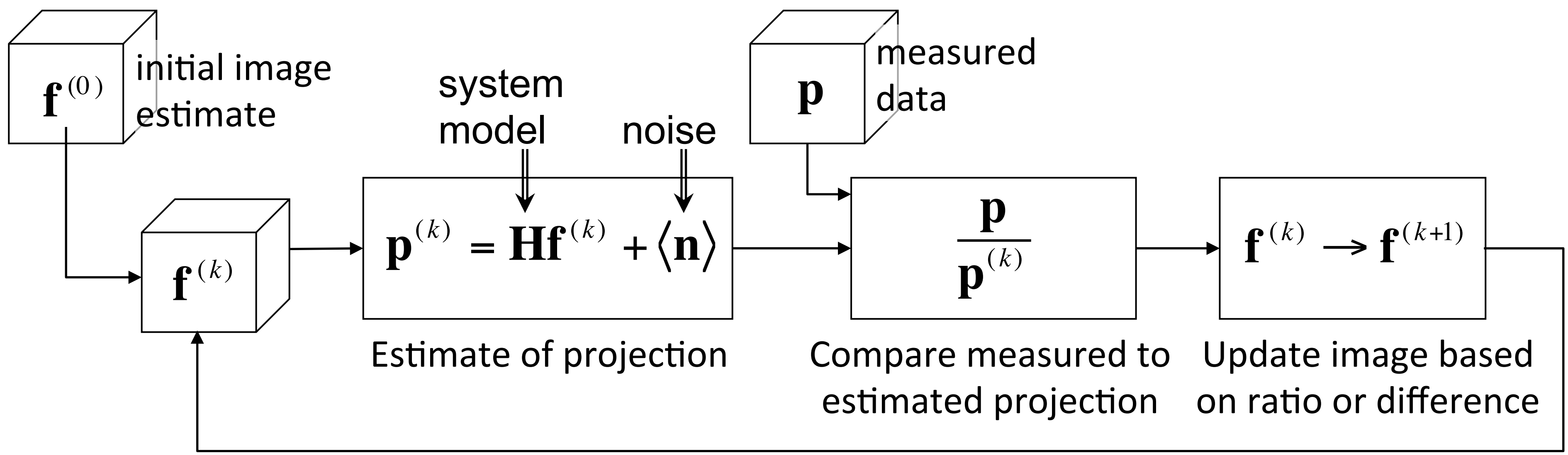


Effect of Butterworth-filter cutoff



Koch W. *J Nucl Med* 2005; 46:1804–11

Iterative reconstruction – a generic example



- Model the system (and the noise) used in projector
- Image update is typically based on differences or ratio of measured and estimated projection data
- Must decide when to stop iterating

Projector can account for

- Signal probability model
- Physics model of SPECT imaging
- System (intrinsic + collimator) spatial resolution
- Patient attenuation
- Compton Scatter (in patient, collimator, crystal)
- Collimator septal penetration

Various Iterative Recon. Methods

- Algebraic reconstruction technique (ART)
- Multiplicative ART (MART)
- Weighted least-squares conjugate gradient (WLS-CG)
- Expectation maximization (EM)

Expectation Maximization (EM)

- Estimates parameters of the statistical distributions underlying the measured data
- Maximize marginal likelihood by iterating two steps:

  Expectation: Marginalize *log likelihood* with respect to the missing data given observed data for the current estimate of parameters

  Maximization: Find set of parameters that maximizes this quantity

- In the case of SPECT
  - Observed data, $\boldsymbol{F}=\{f_j, j=1,\ldots,J)$ are the projections onto detector elements
  - Parameters, $\boldsymbol{A}=\{A_i, i=1,\ldots,I)$ are the true count rate in a voxel at $\{x_i, y_i, z_i\}$
  - The parameters $\{A_i\}$ are independent and Poisson distributed

$$\hat{\boldsymbol{A}} = \underset{\boldsymbol{A}}{\operatorname{argmax}}\{pr(\phi \mid \boldsymbol{A})\} = \prod_{i=1}^{I}\prod_{j=1}^{J}\frac{A_i^{\phi_j}e^{-A_i}}{\phi_j!}$$

Ordered subset EM (OS-EM)

- Performs EM sequentially on non-overlapping subsets of the projection data until all projections are considered
- Must decide # of subsets (n) and # of iterations (m)
- m × n EM iterations, but only m projections of data
- Time for 1 OSEM iteration (all subsets) > 1 EM iteration
- However, OSEM increases convergence rate

Image Acquisition

Number of detectors and orbits

- Single head
  - 180 vs 360 acquisition: speed vs. uniformity tradeoff
  - Conjugate views
  - Non-circular orbits: improved resolution
- Two headed
  - Smaller rotation needed: 360→180 or 180→90
  - H vs. L modes: region of interest and attenuation
- Circular vs. contouring orbits

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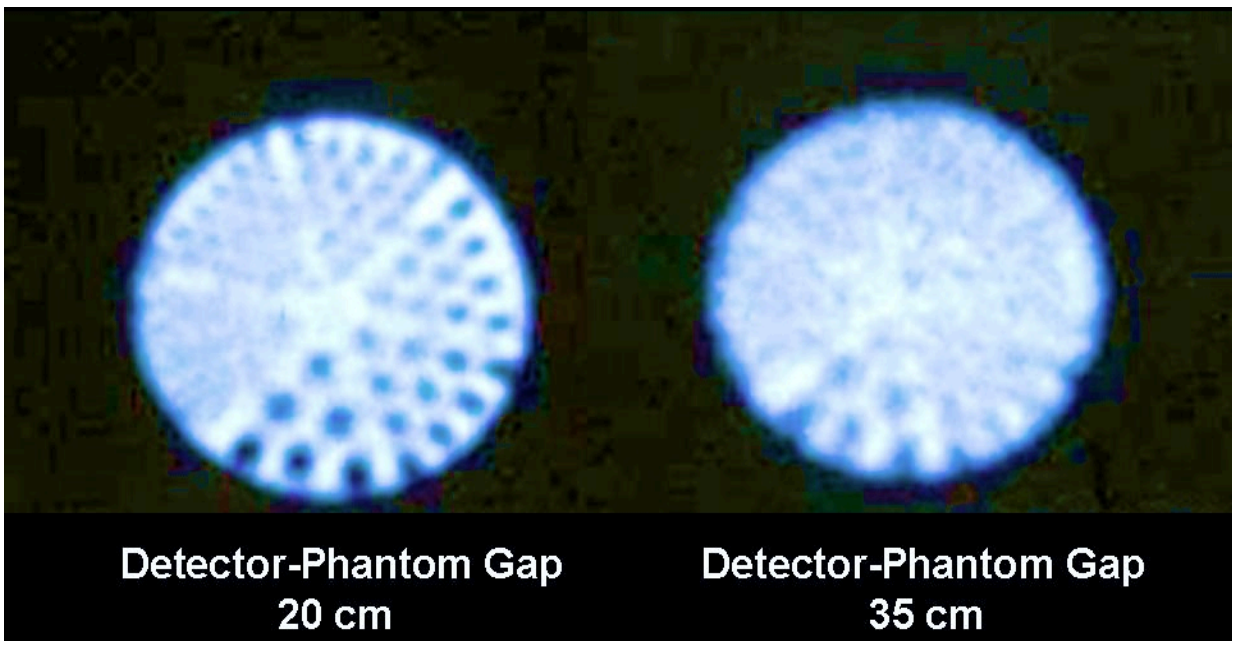
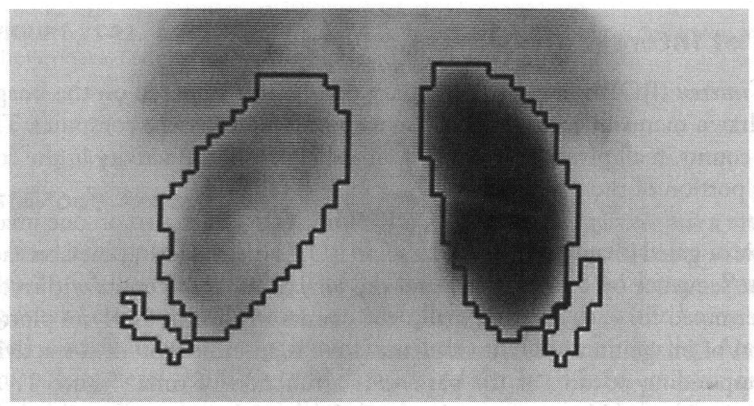


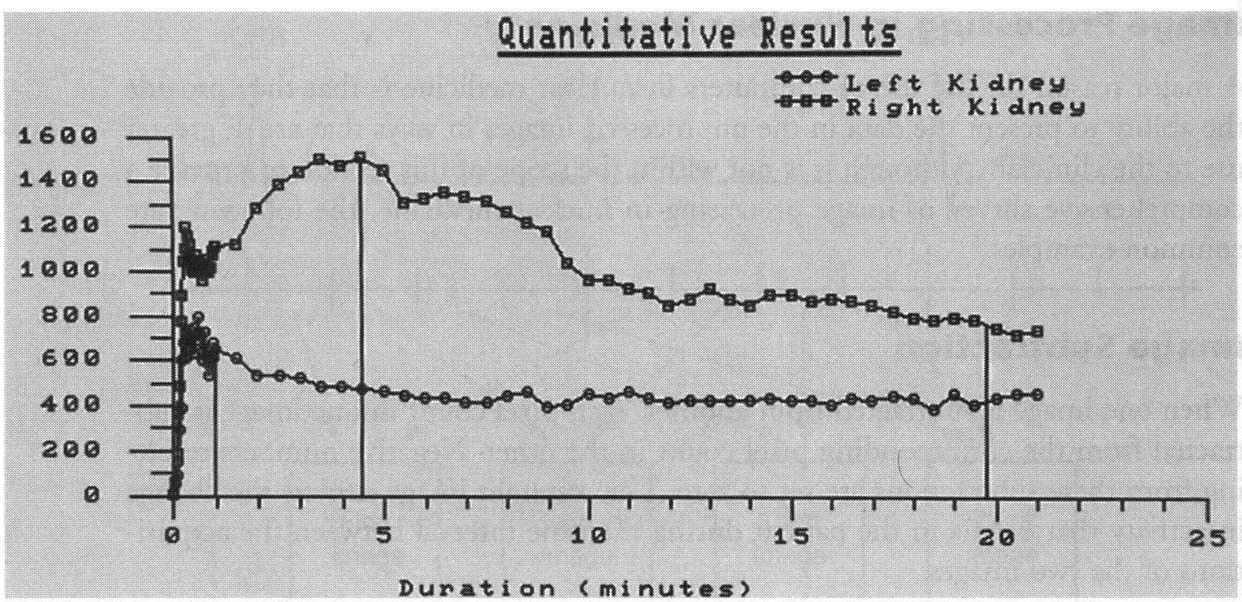
Image Acquisition Modes

- Frame mode (data stored as an image)
  - static
    - single image acquisition
    - can have multiple energy windows
  - dynamic
    - series of images acquired sequentially
  - gated
    - repetitive, dynamic imaging
    - used for cardiac imaging
- List-mode (data stored event by event)
  - time stamps are included within data stream
  - allows for flexible post-acquisition binning
  - can result in very large data files

Dynamic acquisition



Regions of interests (ROIs)



Time-activity curves (TACs)

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

Cardiac Gated Acquisition

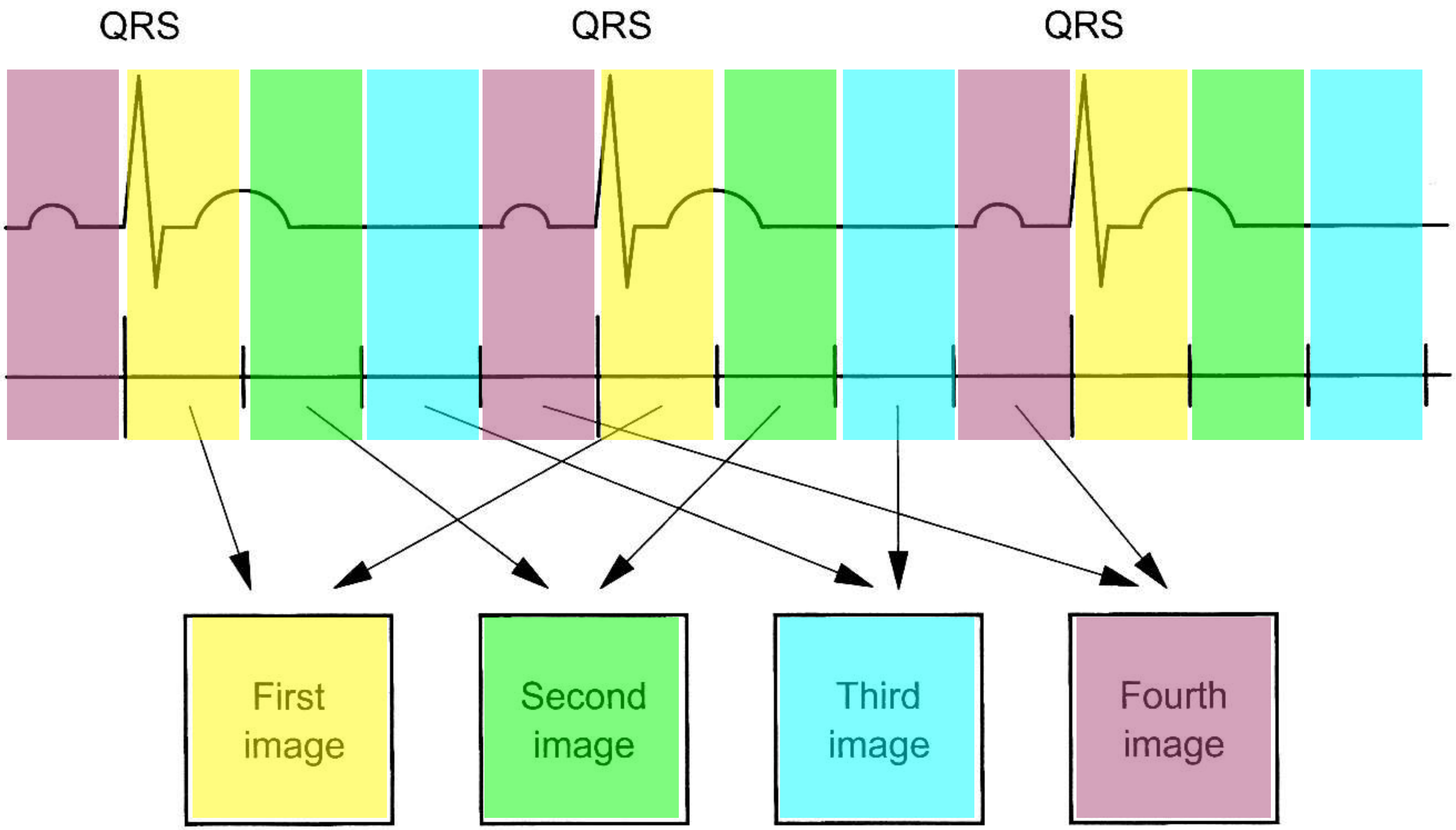
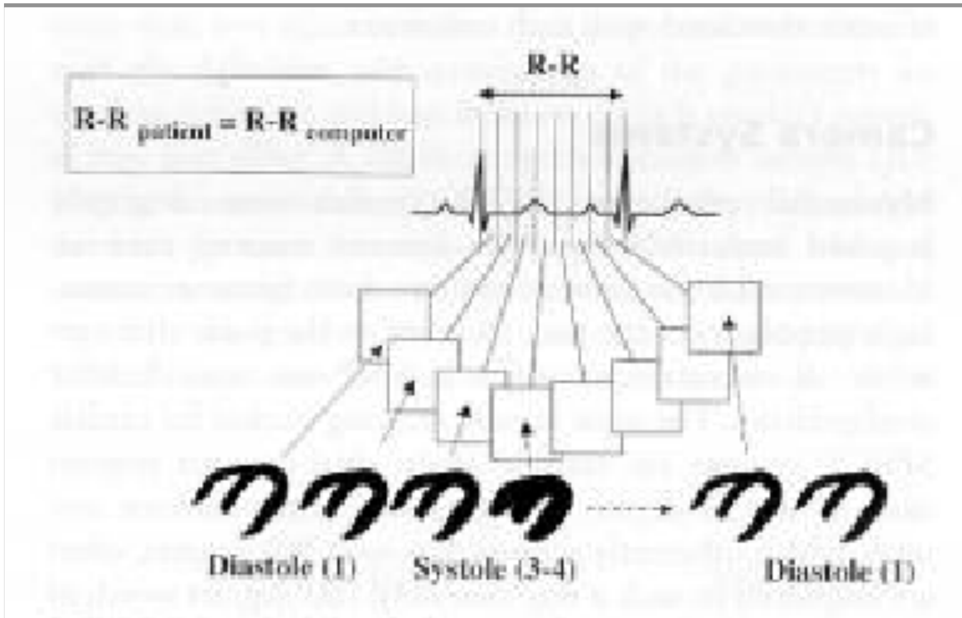
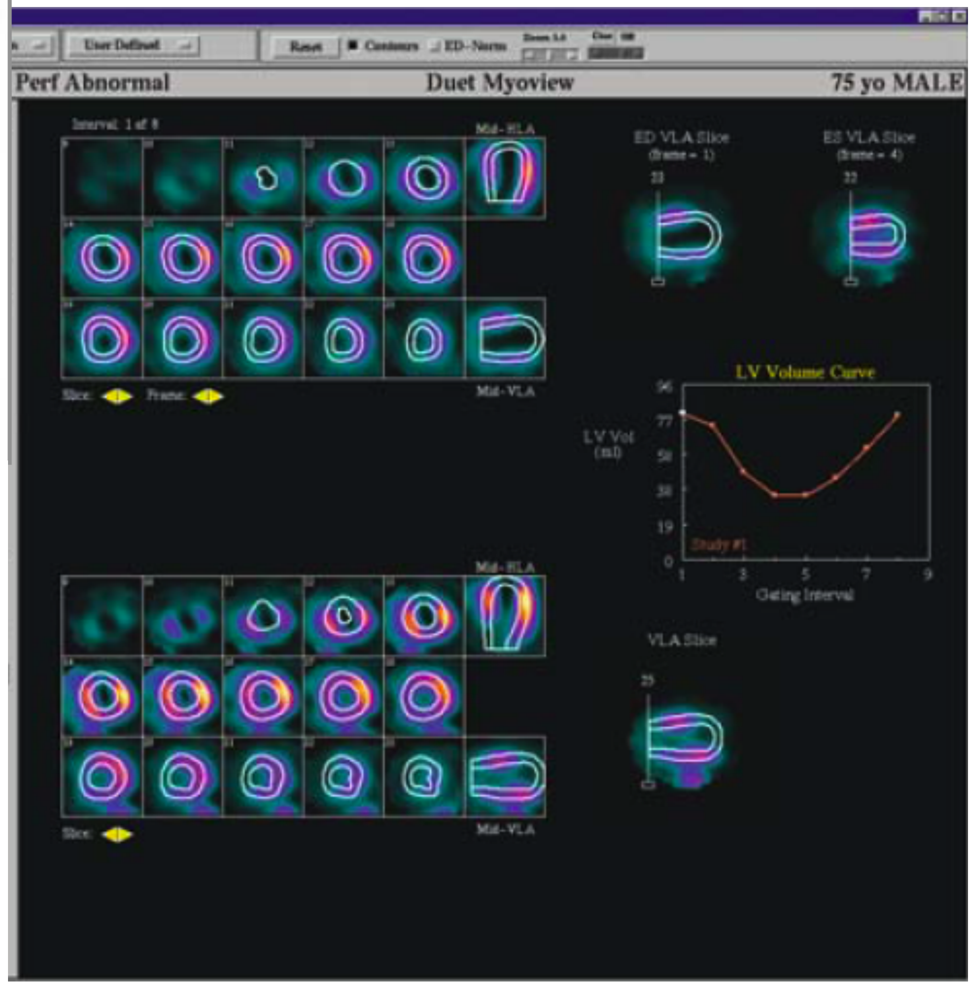


FIGURE 21-22. Acquisition of a gated cardiac image sequence. Only four images are shown here. Sixteen to 24 images are typically acquired.

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

Gated cardiac SPECT

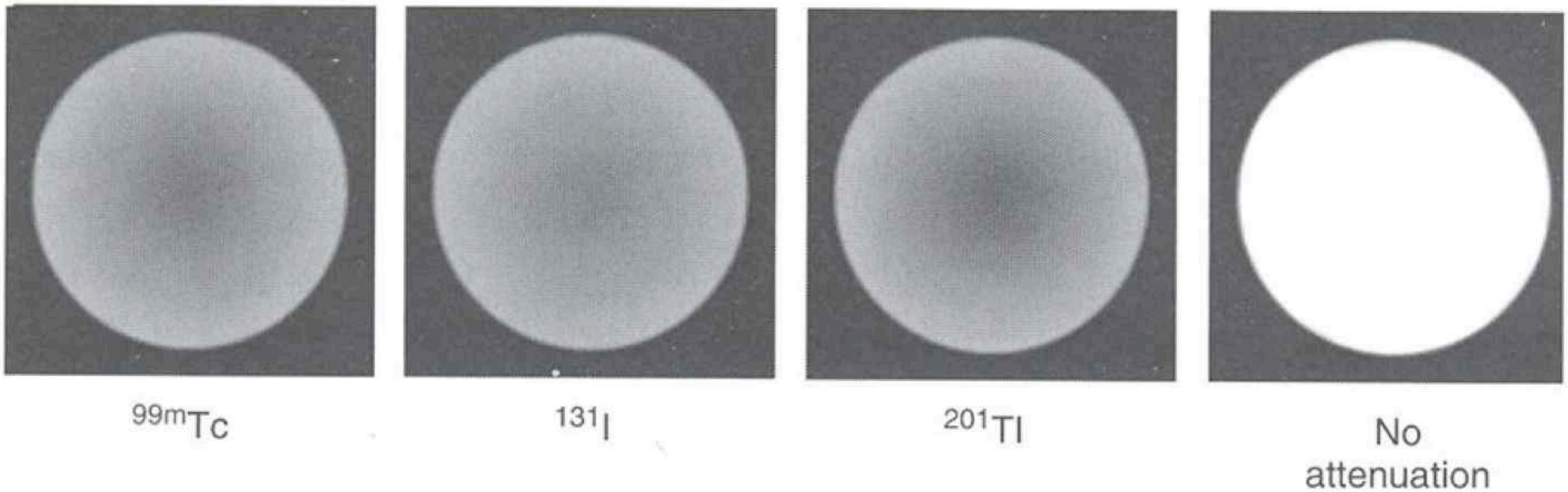




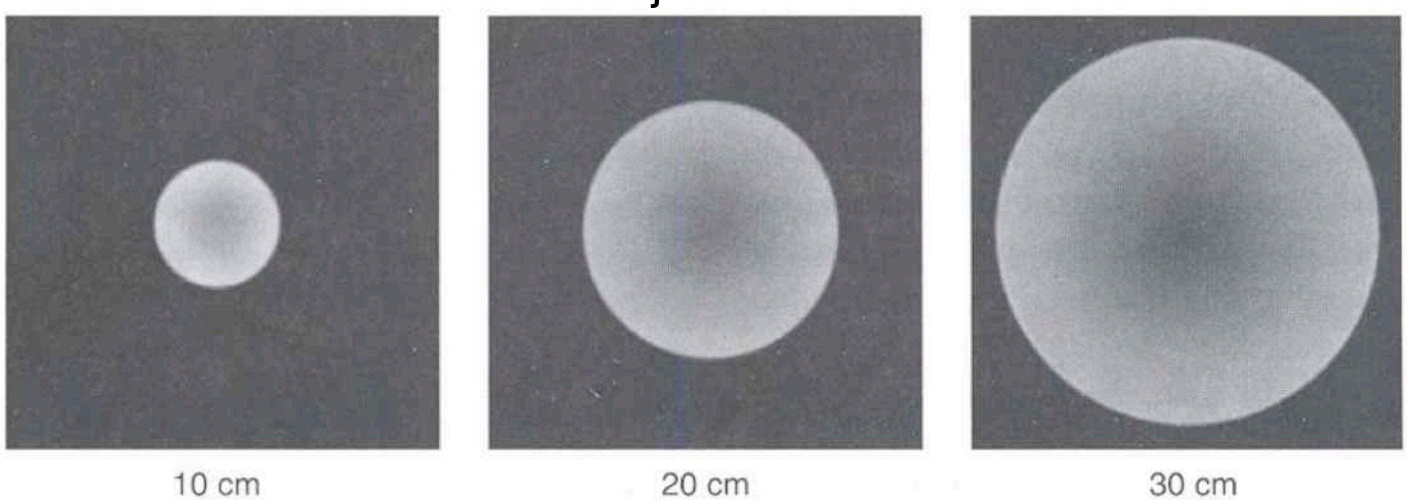
- Images obtained during individual phases of cardiac cycle
- Gating based on EKG
- 8-16 frames per cycle
- Assessment of global and regional (apex, mid, base) left ventricular (LV) function vs. phase (i.e. systolic and diastolic)

Attenuation Effects

Energy dependance



Object size



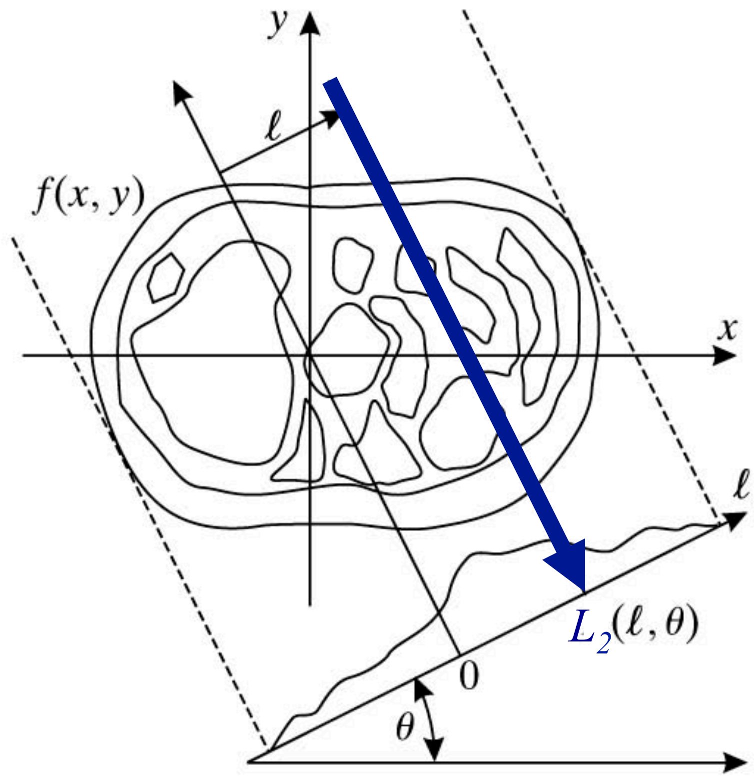
Cherry SR, In: *Physics in Nuclear Medicine*, 2003, p 308 & 311

Methods of Attenuation Correction

- Transmission measurement
  - Scanned line source
  - Moving energy window: different energy for transmission & emission photons
  - Low signal-to-noise due to practical transmission activity
  - Direct measure of attenuation coefficient

Methods of Attenuation Correction

- Conjugate views
  - Arithmetic mean: $(L_1+L_2)/2$
  - Geometric mean: $\sqrt{}(L_1^2+L_2^2)$



Methods of Attenuation Correction

- Chang correction
  - Assumes constant μ within anatomical boundary
  - Requires accurate anatomical boundary definition
  - Empirically adjusted for out-of-plane scatter
  - Accurate boundaries difficult to obtain

Where will this work well?

Brain SPECT

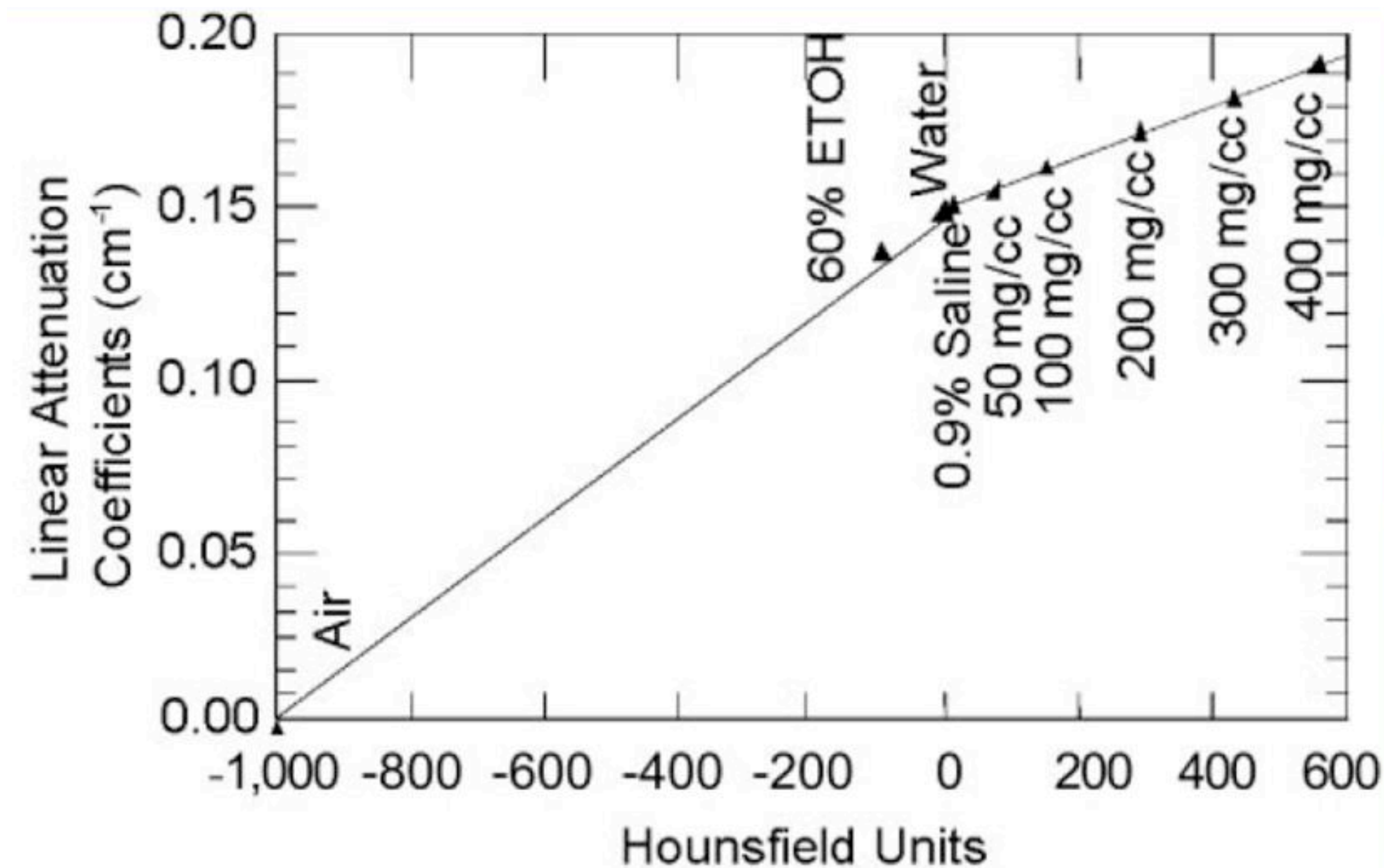
Where is this problematic?

Variable μ values (esp. in thorax)

Methods of Attenuation Correction

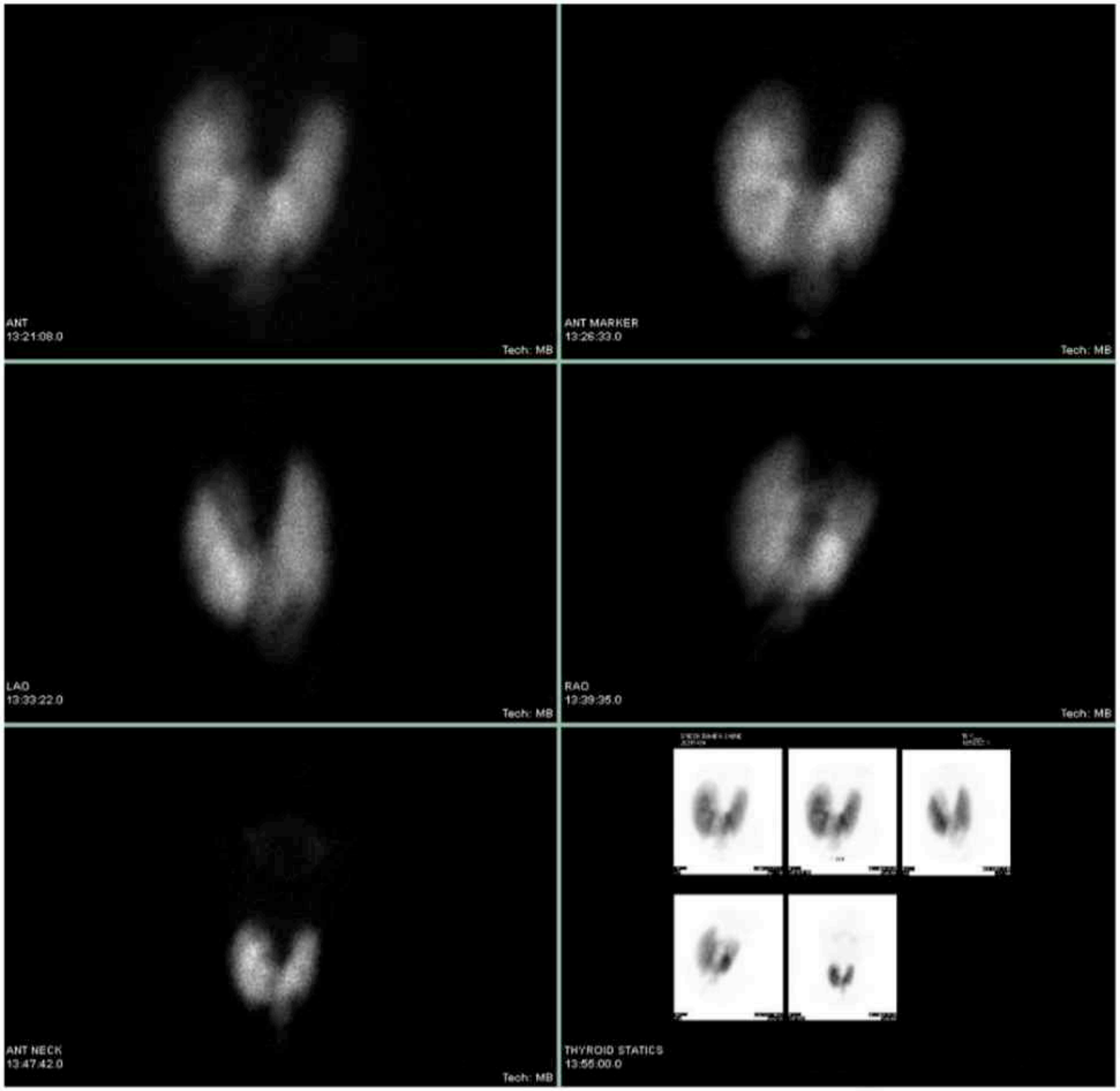
- X-ray CT
  - Co-located and co-registered
  - Lower energy than transmission delivers lower dose
  - Conversion from CT Hounsfield units to equivalent linear attenuation coefficient

Converting Hounsfield Units to attenuation coefficients



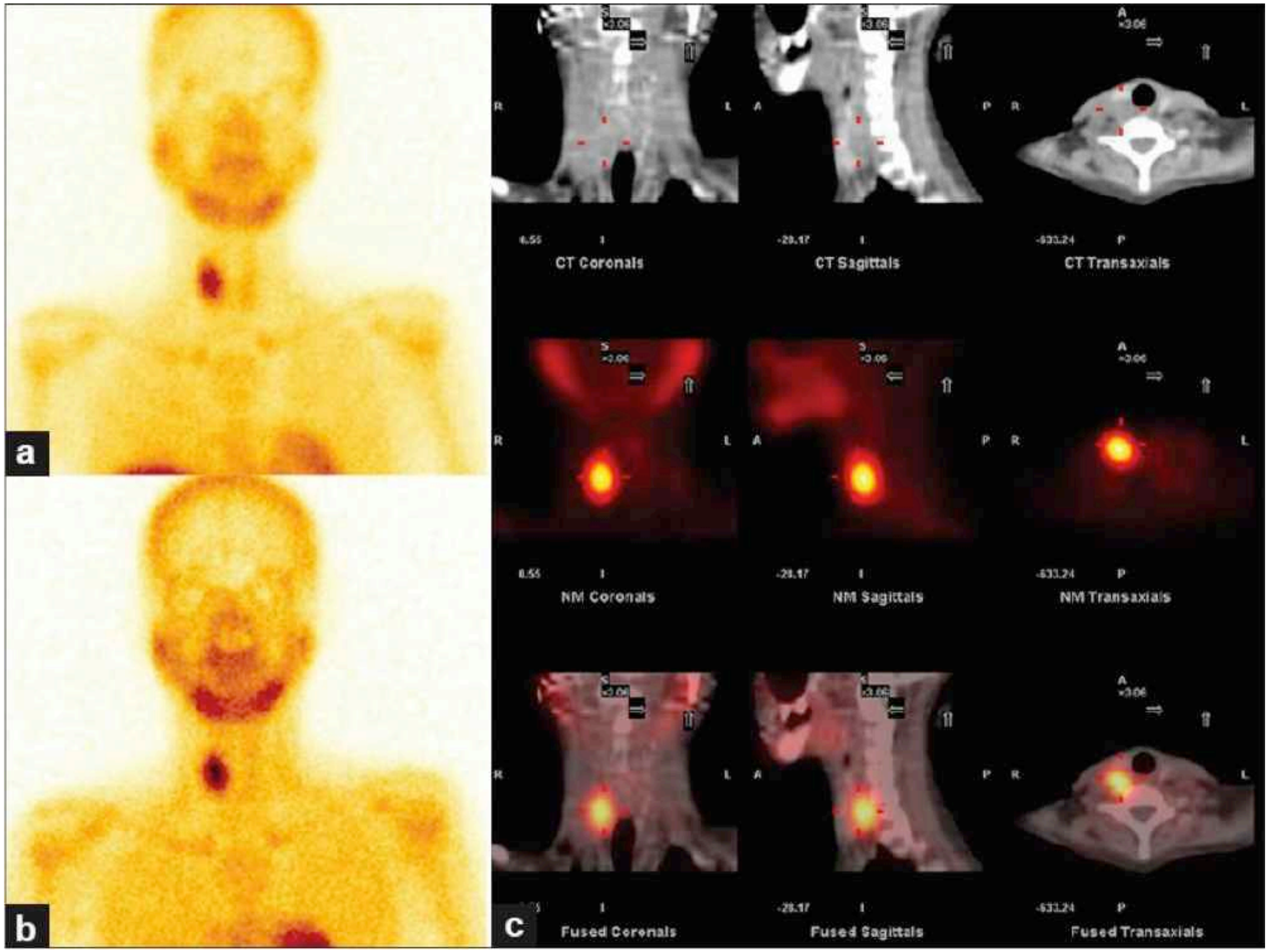
Examples of SPECT imaging

Thyroid Imaging



Sample nuclear medicine thyroid images. The main characteristics used for interpretation of the images are size of the thyroids and whether there is uniform uptake between the left and right thyroids.

SPECT/CT Thyroid Image Example



A 37 YOF with elevated Ca, decreased phosphorus and increased parathyroid hormone

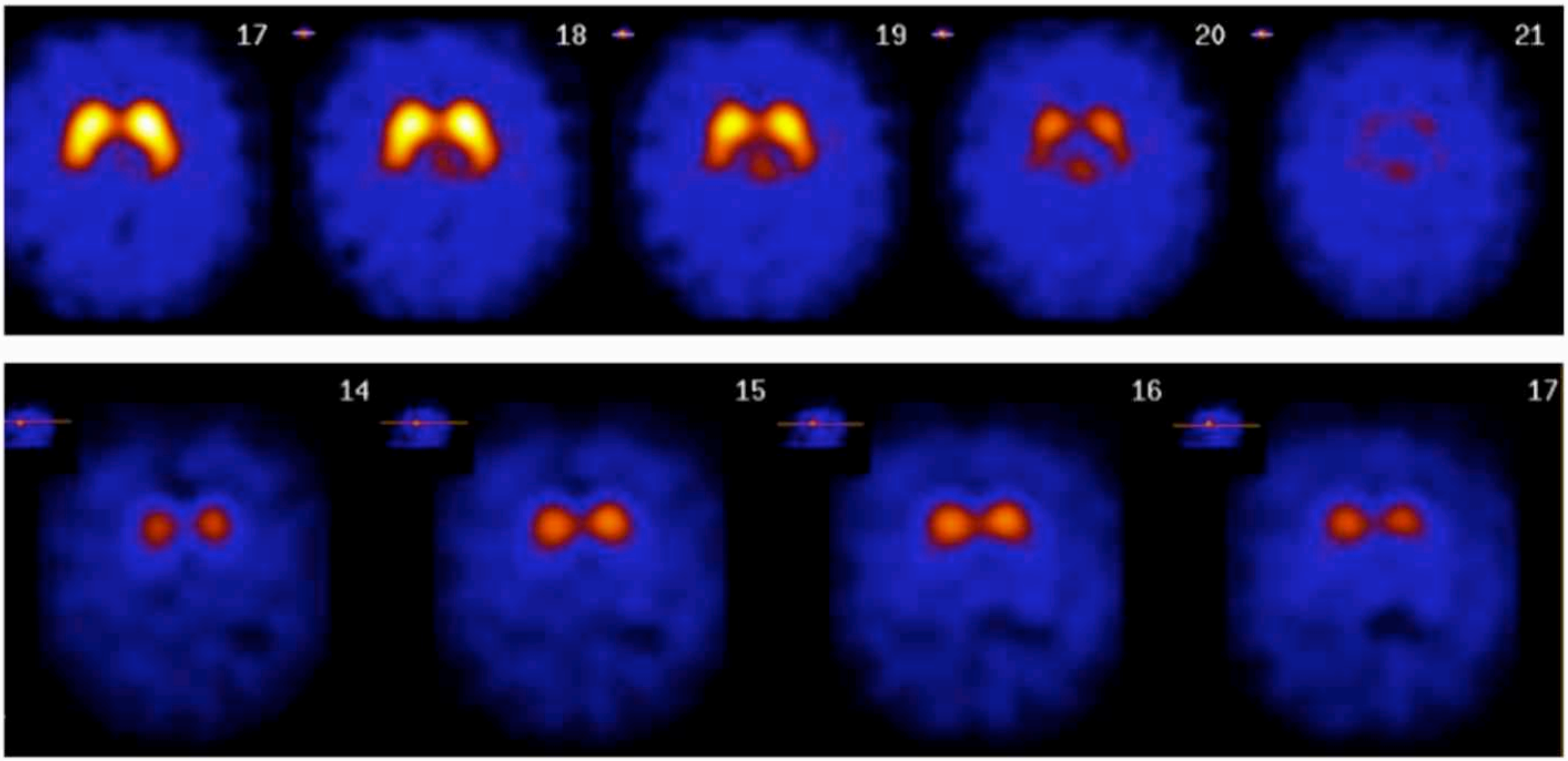
(a), (b) early and delayed Tc99m-sestamibi scintigraphs showing uptake near thyroid

(c) CT, SPECT, and fused SPECT/CT images localize increased tracer uptake to the posterior aspect of right lobe of the thyroid confirming a right superior parathyroid adenoma

Renal Imaging

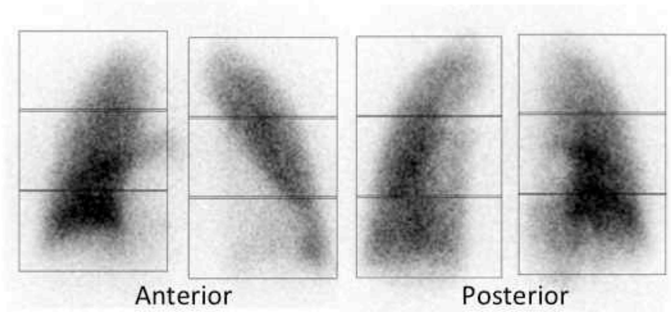
A renogram provides a time-activity curve of the uptake and excretion of a radiotracer by the kidneys. It is used to both evaluate renal function and if there are any bilateral differences between the kidneys. There is often a perfusion phase and a functional phase of the exam. A standard protocol is 80 one second frames to visualize kidney perfusion and 120 twenty second frames to evaluate function.

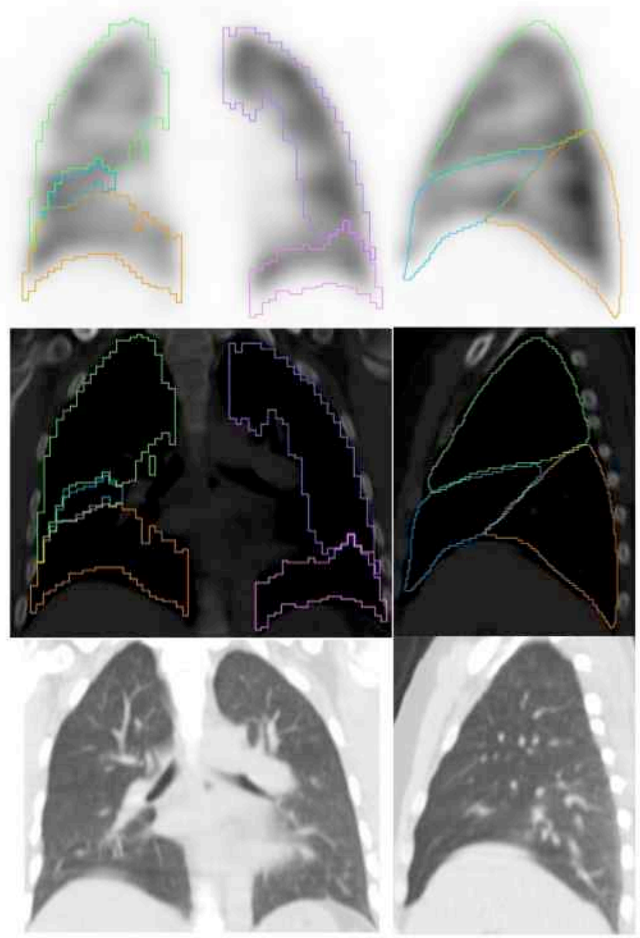
Brain Scan



In this example, an imaging agent called DaTscan is used to differentiate between Parkinsonian syndromes (PS) and essential tremor. The “comma” image is for a patient without PS (upper panel), while the abnormal “period” scan is for a patient with PS (lower panel). DaTscan is a radiopharmaceutical indicated for striatal dopamine transporter visualization.

SPECT/CT Lung Perfusion Imaging





Lung zone regional perfusion estimated by geometric mean of planar counts			Lobar quantitative perfusion by attenuation-corrected SPECT/CT		
ZONES	Right	Left	**LOBES**	Right	Left
Upper 1/3	8.8%	10.4%	Upper Lobe	22.5%	24.8%
Middle 1/3	24.1%	18.9%	Middle Lobe	8.8%	-
Lower 1/3	22.4%	15.0 %	Lower Lobe	23.1	20.9%
Total Lung	**55.3%**	**44.7%**	**Total Lung**	**54.3%**	**45.7%**

Tc-99 MAA lung perfusion with planar perfusion (upper left) and SPECT/CT with attenuation correction (upper right). The estimated perfusion contribution of each lung obtained by both methods is nearly identical (lower left). However, the SPECT/CT method provides more anatomically accurate lobar perfusion quantitation.