Phys. 428

LECTURE	DATE	INSTRUCTOR	TOPIC
1	April 2	PK	Overview: Imaging equation, inverse problem
2	April 9	PK	2D-LSI imaging systems, X-ray physics: formation and interaction
3	April 16	WH	X-ray detection and imaging systems
4	April 23	PK	X-ray computed tomography (CT) systems
5	April 30	WH	X-ray CT part 2. Contrast Agents
6	May 7	PK	Image reconstruction and image quality
7	May 14	LM	Nuclear decay schemes and isotopes
8	May 21	RM	Gamma cameras: components and systems
9	May 28	WH	Tomography in molecular imaging: SPECT scanners
10	June 4	SB	Positron emission tomography (PET) and hybrid PET/CT scanners
11	June 13	WH/PK	Group project presentations **

*** note change in date for class presentations to June 13th - Thursday instead of Tuesday*

Each student should email at least one question on today's lecture to our TA, Jackie (jackie24@uw.edu) by Friday.

Please include "Phys 428 Lecture 9 Question" in the subject line.

Single Photon Emission Computed Tomography (SPECT) Imaging

Aim:

To map physiological processes in vivo
using single-photon-emitting tracer compounds

Method:

Label compound with radioactive isotope.
Acquire projection images of emitted radiation.
Reconstruct tracer distribution from multiple projections.

Analogies with X-ray Imaging

	X-rays	**Nuclear medicine**
Photon formation:	bremsstrahlung characteristic x-rays	positron emission isomeric transition
Attenuation by tissues:	photoelectric absorption Compton scatter	photoelectric absorption Compton scatter
Detection systems:	Scintillators	Scintillators
2-D projection imaging:	radiographs	scintigraphs
Tomographic imaging:	CT collect projections from around the patient	SPECT collect projections from around the patient

Difference with X-ray Imaging

X-rays

Transmission

Nuclear medicine

Emission

What is the primary difference of these two modalities?

Gamma Camera Imaging (SPECT)

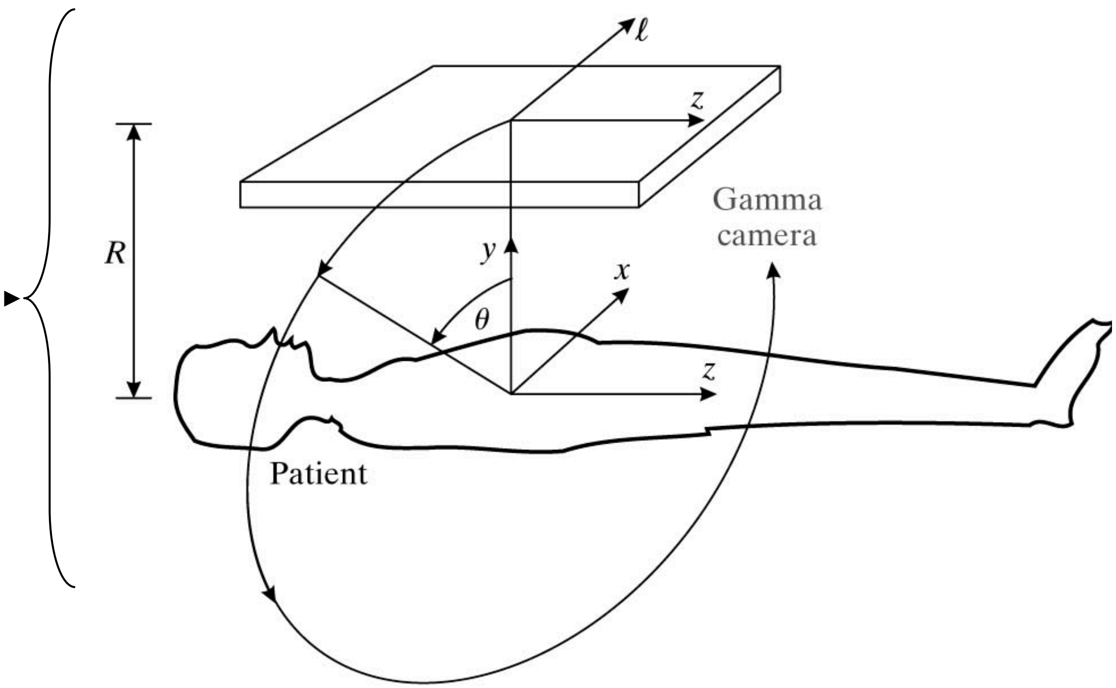
Siemens Healthcare, Symbia

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



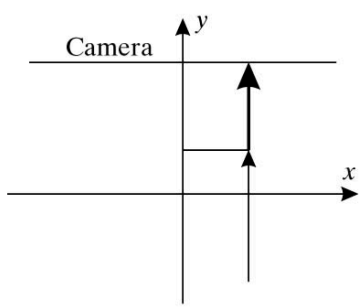
SPECT Imaging Equation

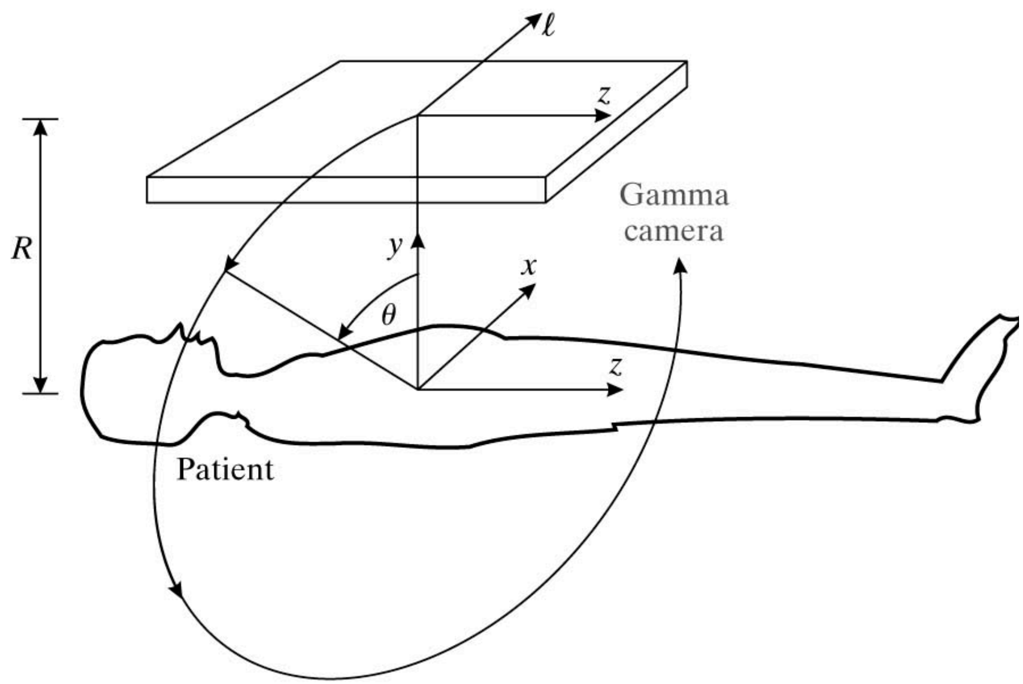
- Using parallel-hole collimators, and ignoring depth-dependent collimator burring etc., we have

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

- 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
    - Natterer F., Inversion of the attenuated Radon Transform, Inverse Problems, vol. 17, no. 1, pp.113 -119 2001
    - Novikov, An inversion formula for the attenuated X-ray transform, Preprint, Dept. de Math, Nantes, April 2000

integration geometry

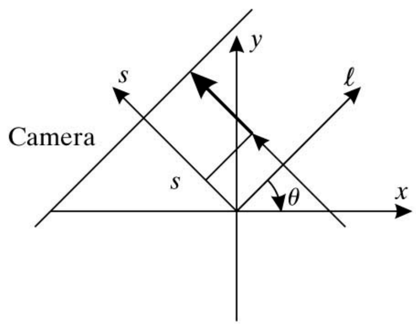




SPECT Image Reconstruction

- 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

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

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

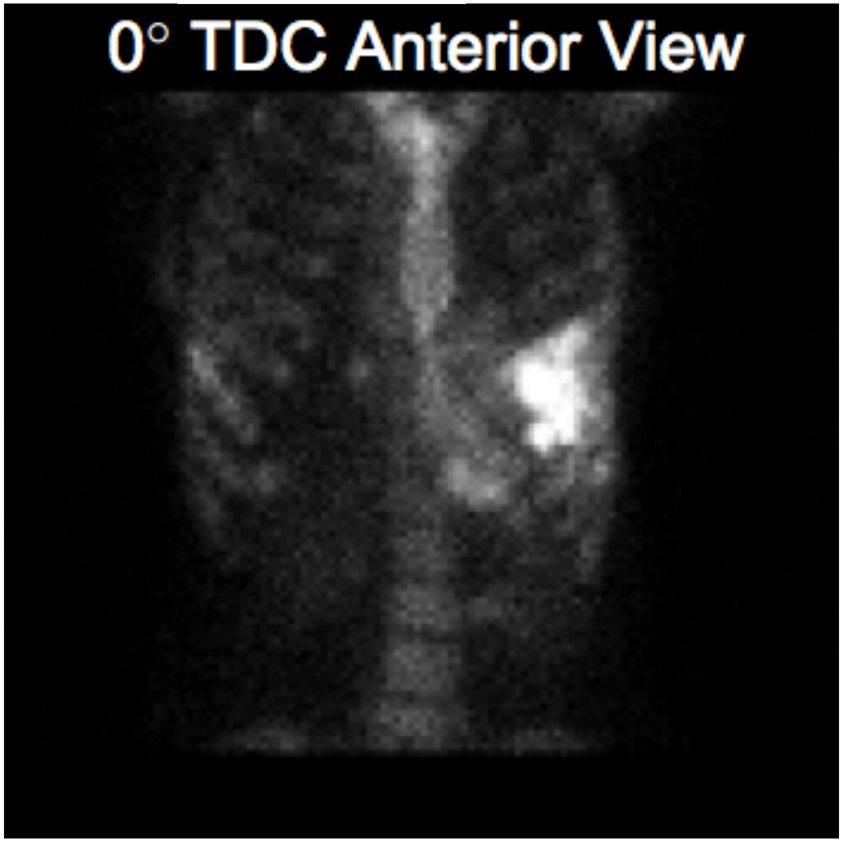
which is the 2-D x-ray or Radon transform that we can solve with FBP

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

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

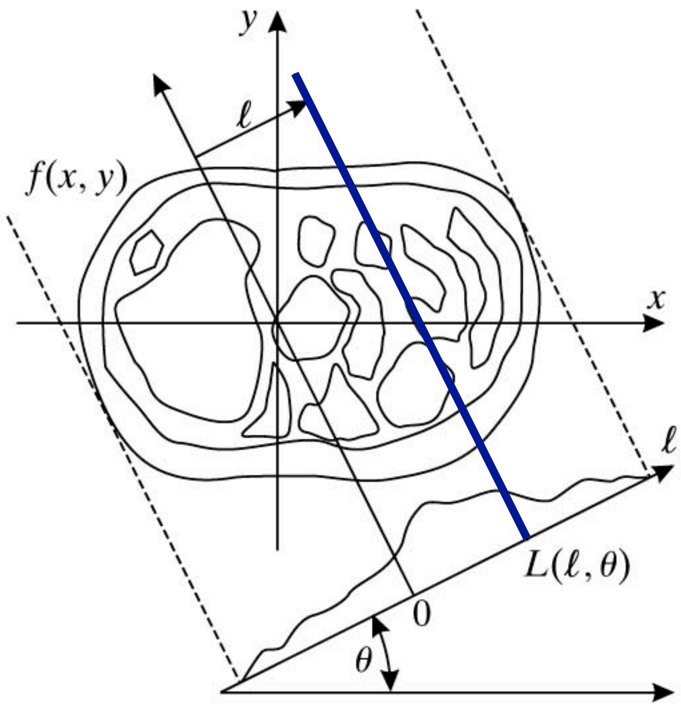
Radon transform angular symmetry violated

$\phi(l,\theta)$



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

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

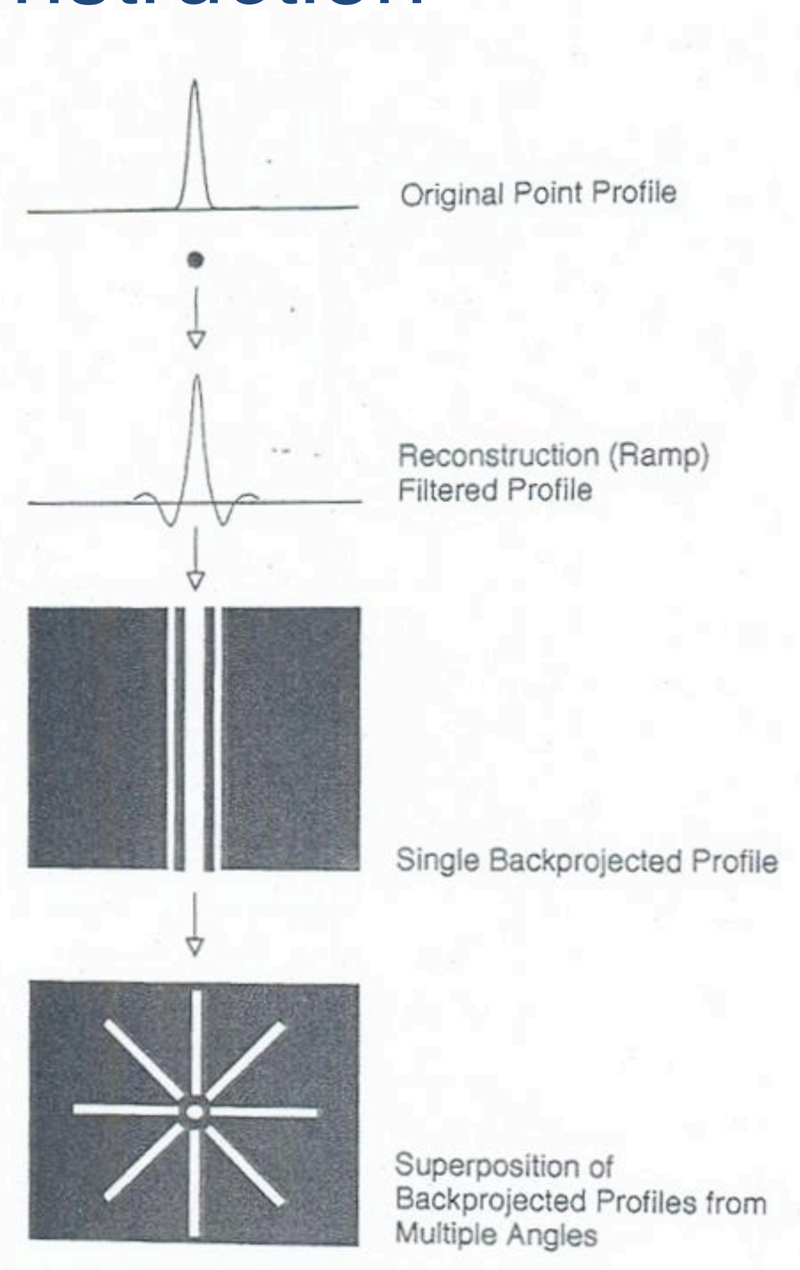
Image reconstruction

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.

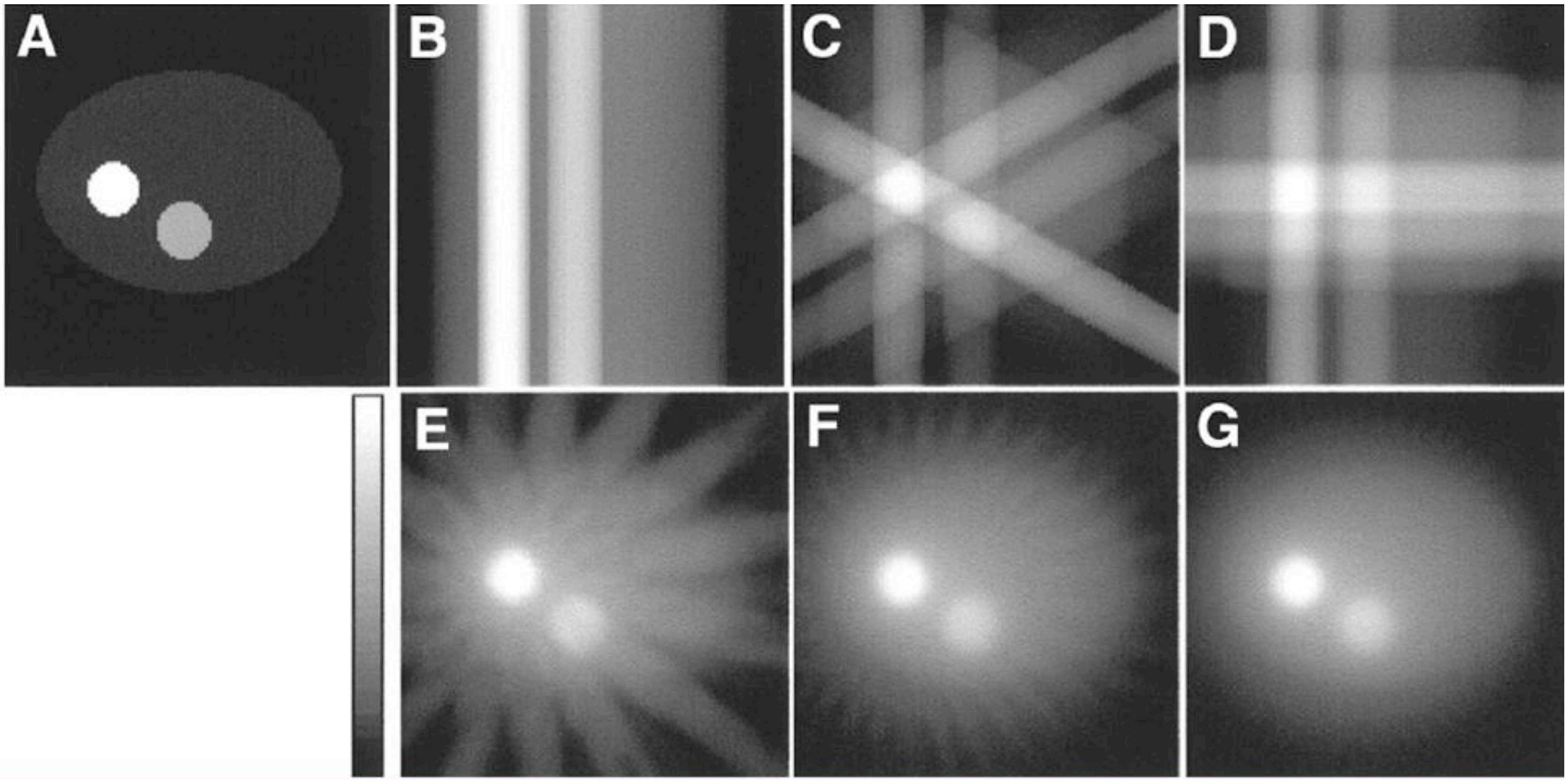
The back projections from different angles are added together to form the reconstructed image



Cullom SJ, In: Cardiac SPECT Imaging, 2001

Image reconstruction

Star (or streak) artifact of FBP reconstruction



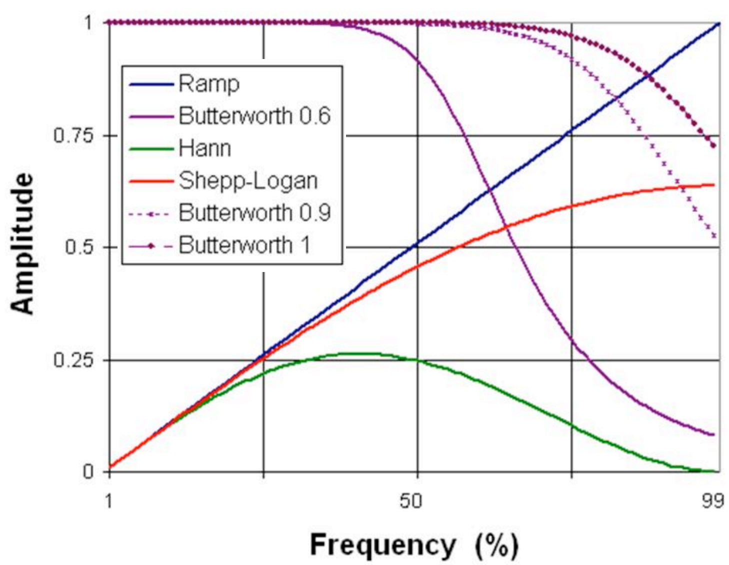
- A = object
- B through G = {1, 3, 4, 16, 32, 64} number of projections
- Star artifact decreases with number of projections

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

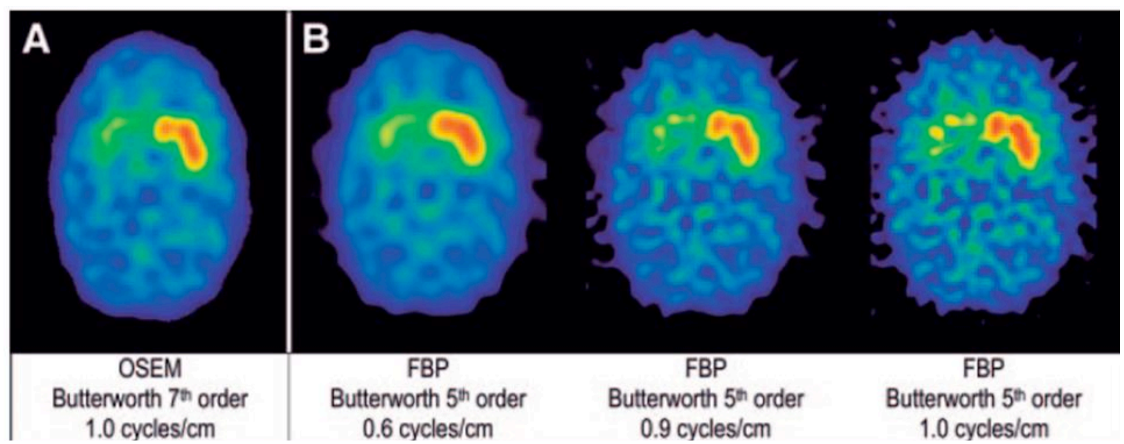
Reconstruction Filters
used to reduce image noise

- 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

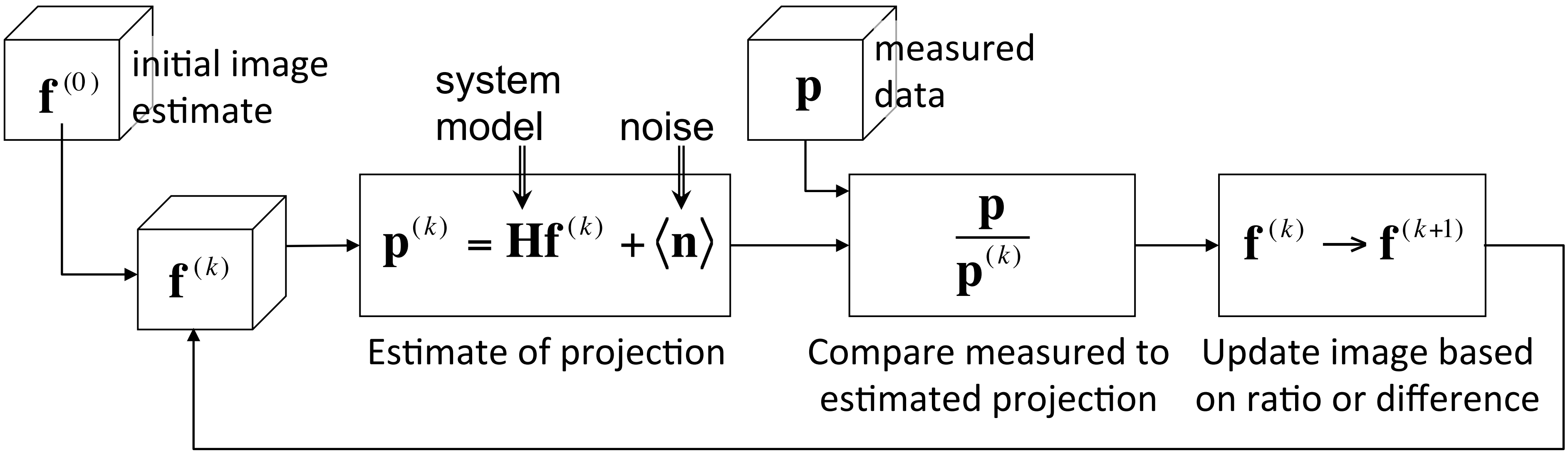


Effect of Butterworth-filter cutoff



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, $\boldsymbol{\Phi} = \{\phi_j\ ,\ j=1,\ldots,J)$ 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)

- $m \times n$ 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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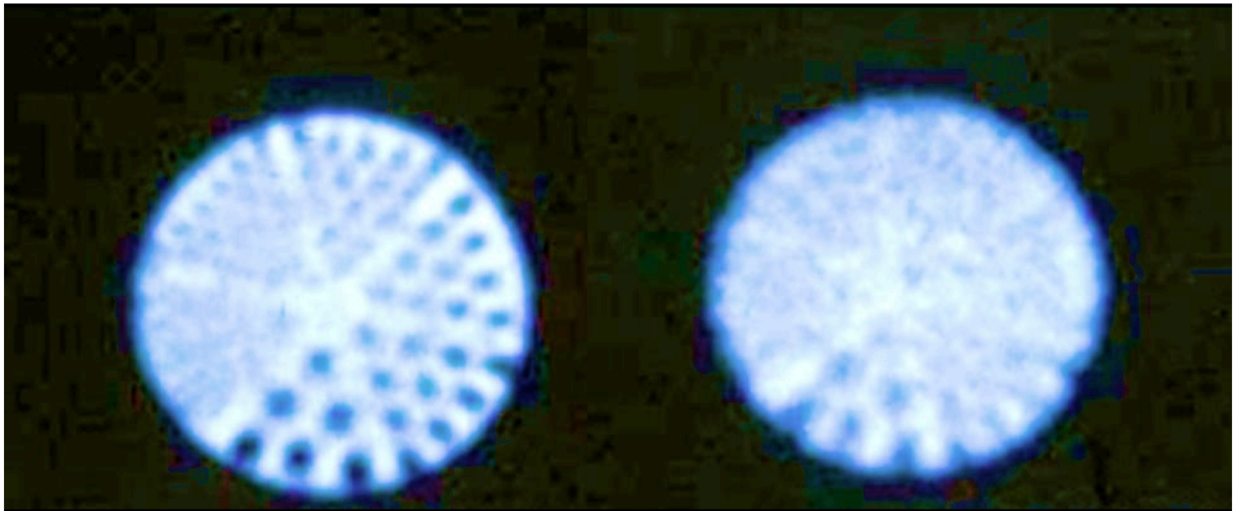


Detector-Phantom Gap
20 cm

Detector-Phantom Gap
35 cm

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

Gated Acquisition

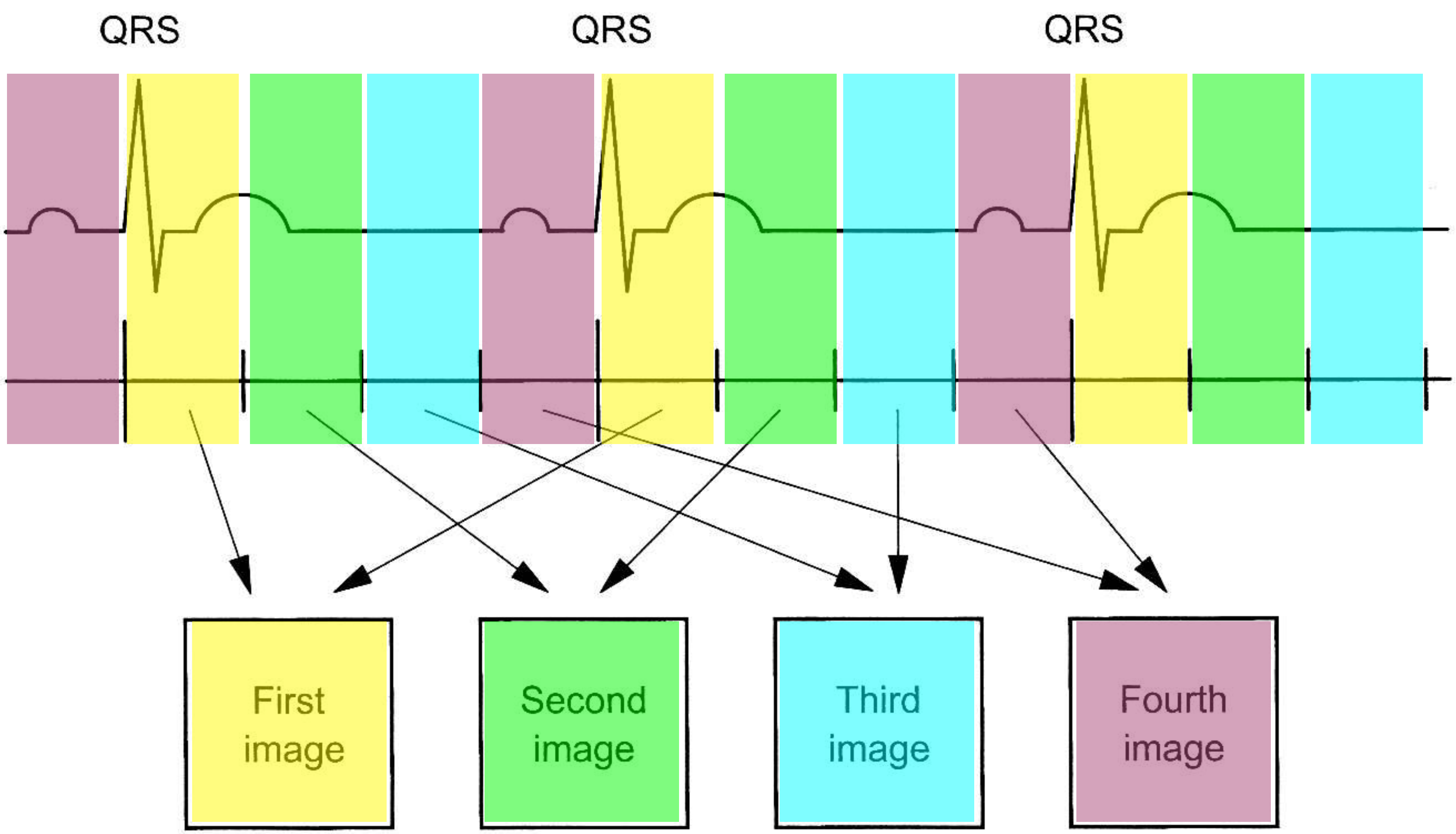
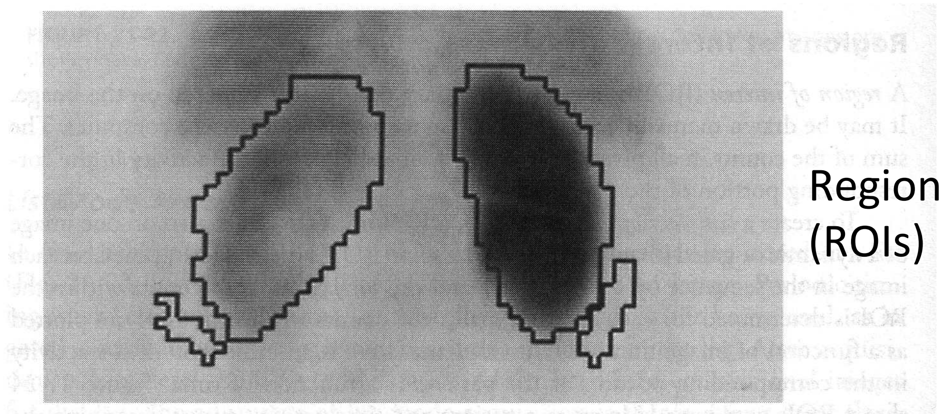


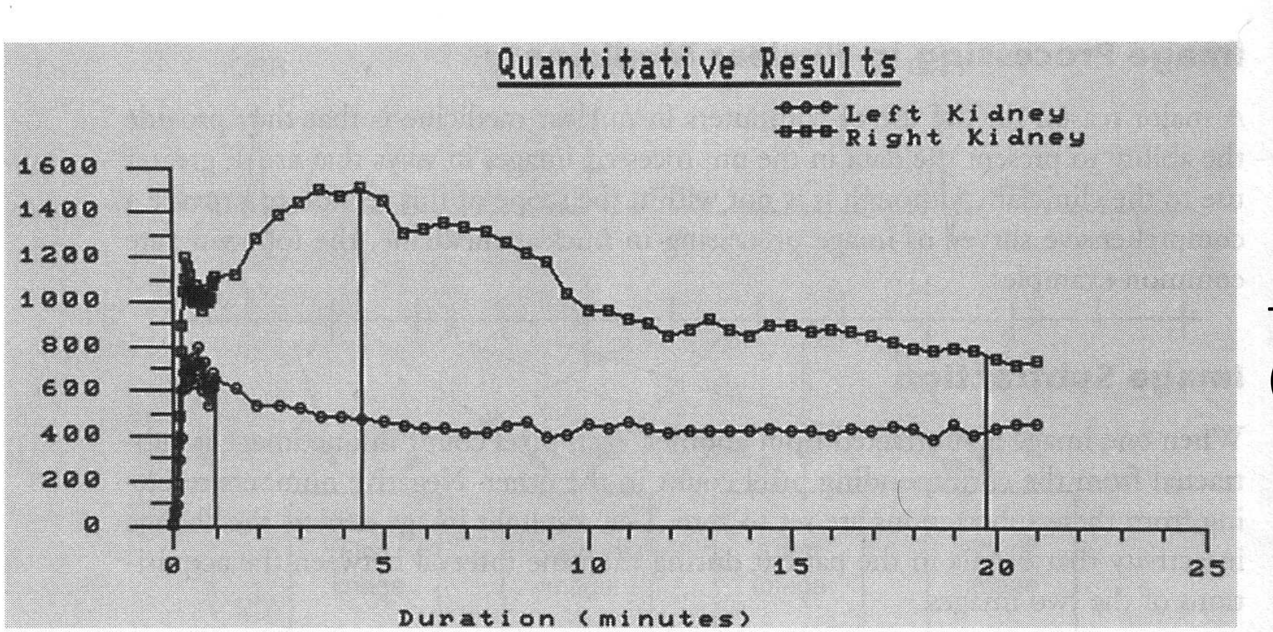
FIGURE 21-22. Acquisition of a gated cardiac image sequence. Only four images are shown here. Sixteen to 24 images are typically acquired.

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

Dynamic acquisition



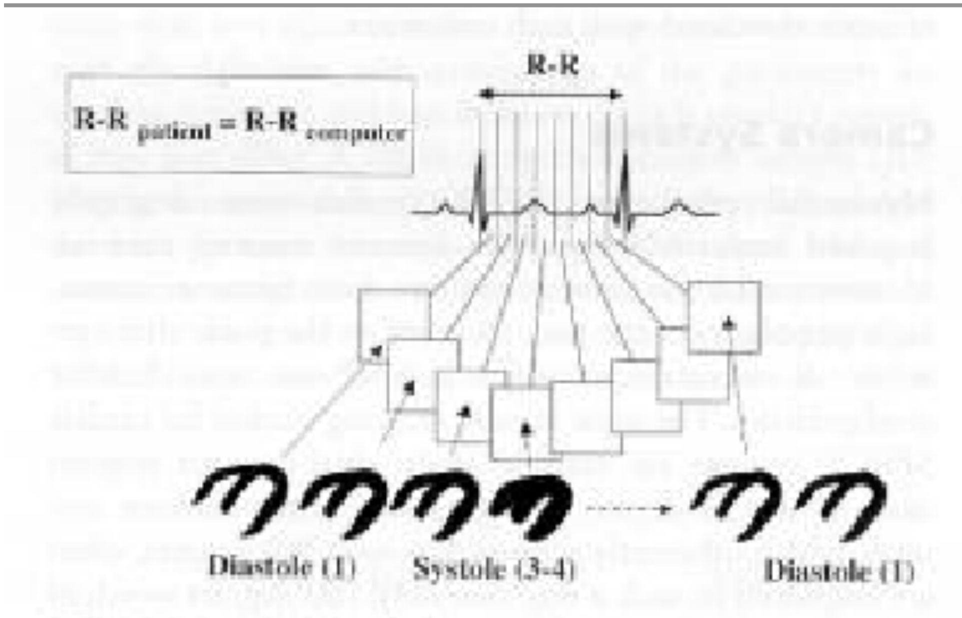
Regions of interests (ROIs)

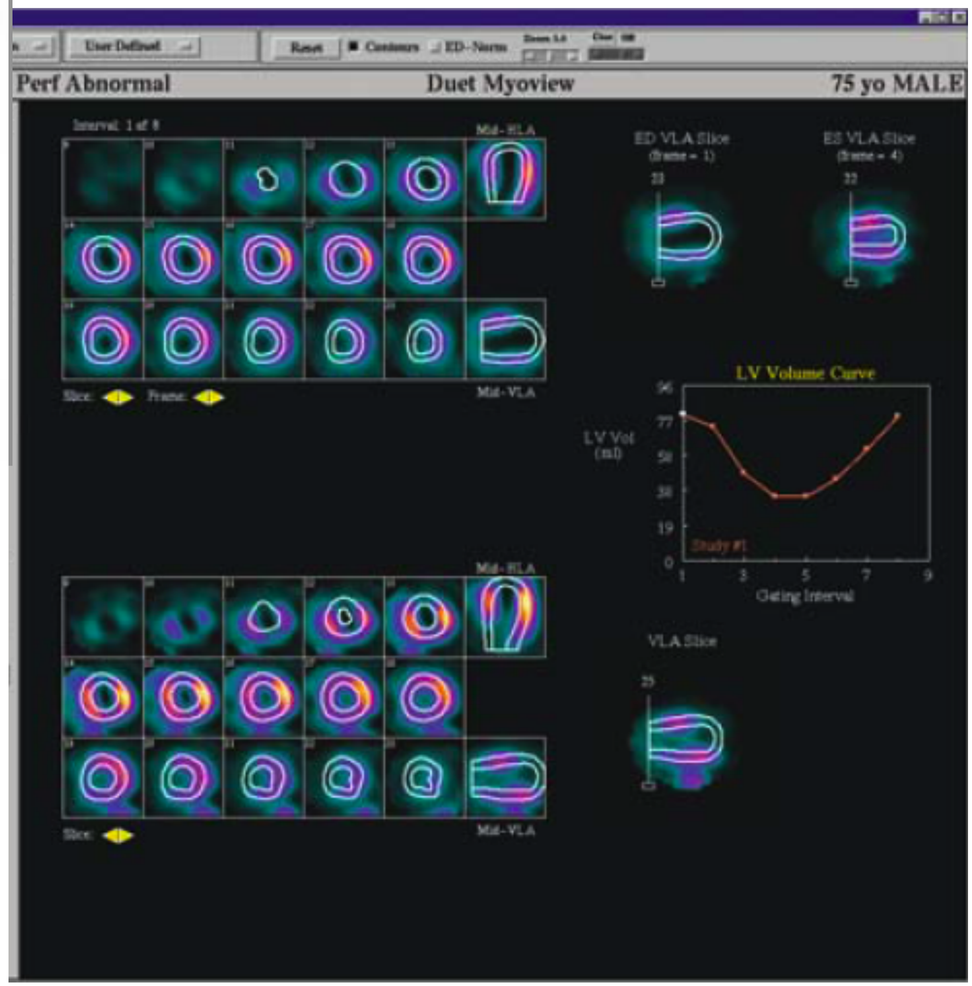


Time-activity curves (TACs)

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

Gated cardiac SPECT





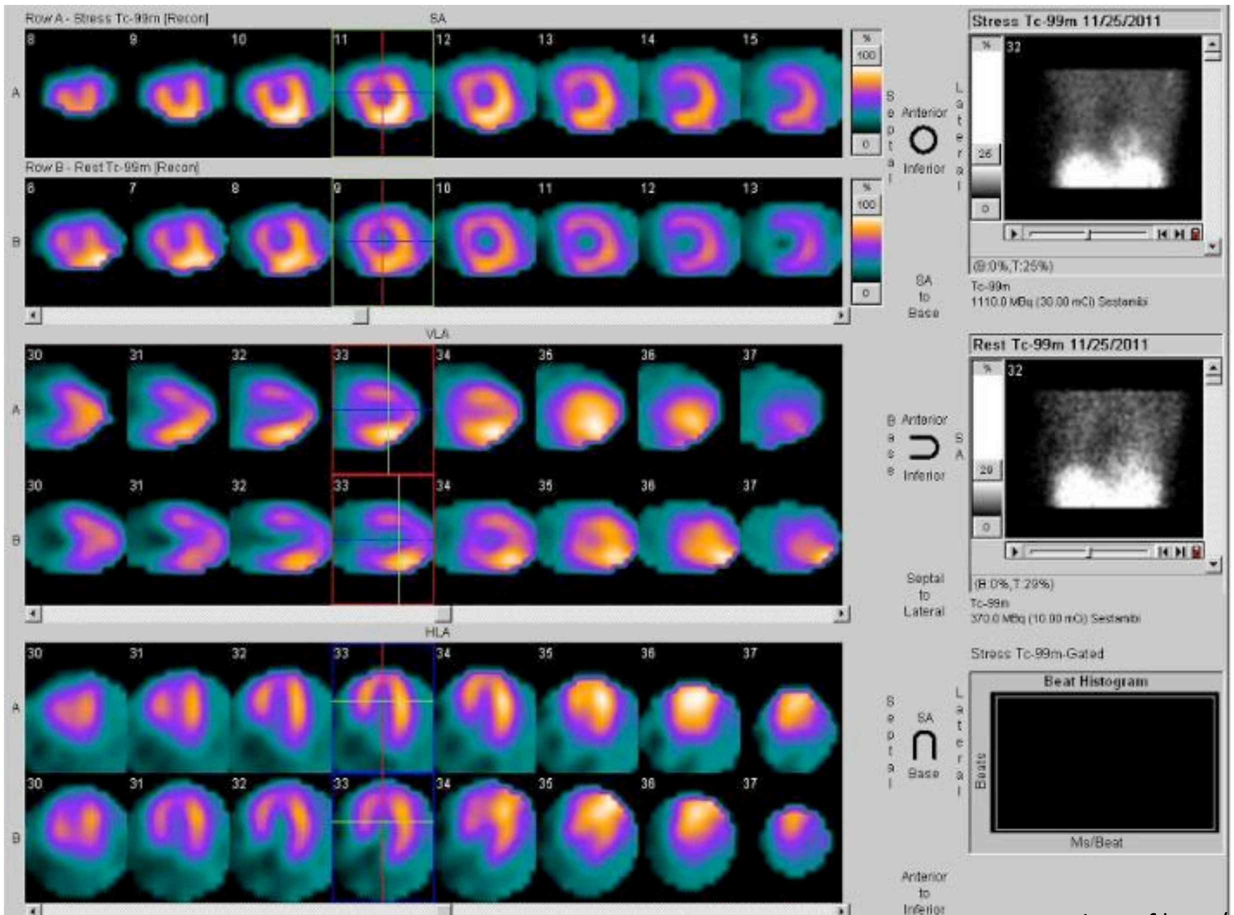
- Images obtained during individual phases of cardiac cycle
- Gating based on EKG
- 8-16 frames per cycle
- Assessment of global and regional (apex, mid, base) left ventricular (LV) function vs. phase (i.e. systolic and diastolic)

www.medical.siemens.com/siemens/en_INT/ gg_nm_FBAs/files/ brochures/Cardiology_Insert_4DMSPECT.pdf

www.cardiology.utmb.edu/ slides/Gated%20Cardiac%20SPECT.htm

NM Physics Theme Example

A 64 yr old female is presented with chest pain and dyspnea. Myocardial perfusion imaging was performed using a one day protocol and exercise stress. She exercised on treadmill following Bruce protocol for 3:46 minutes. Resting heart rate of 66 beats per minute rose to 158 beats per minute, 87% predicted maximum heart rate. She received 0.37 GBq (10.0 mCi) Tc-99m Sestamibi IV at rest and 1.11 GBq (30.0 mCi) Tc-99m Sestamibi IV at stress.



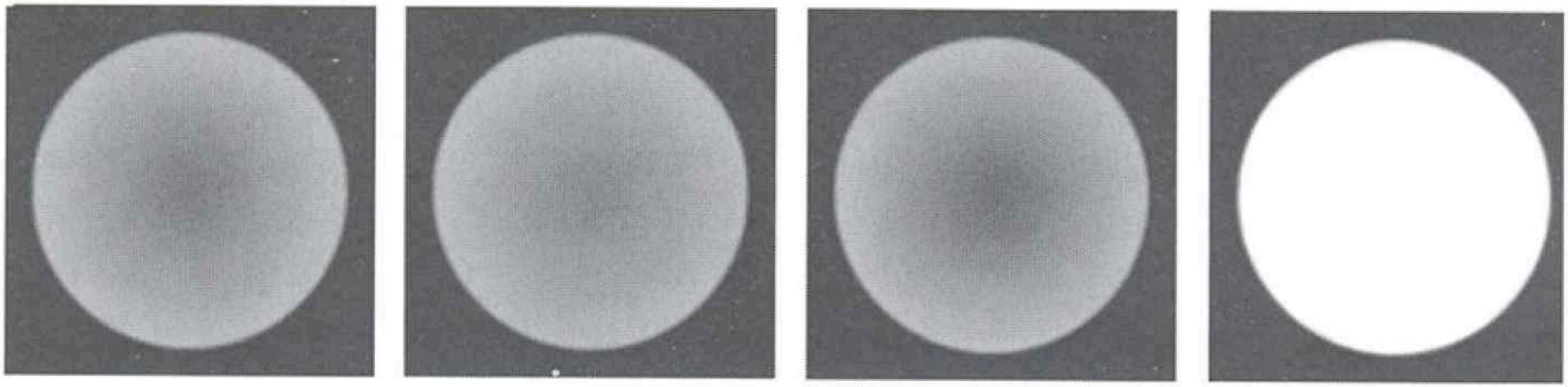
How do you interpret these findings?

The images do not show any reversible perfusion defects to suggest exercise induced myocardial ischemia. There is a moderately severe, fixed, myocardial perfusion defect involving the anterior wall of the left ventricle, worse at rest. These findings are characteristic of a breast implant attenuation artifact. The MIP images show a photopenic implant in the left breast, in the left anterior oblique projection.

variant of http://www.nuclearmd.com/Nuclear-Cardiology-Case-405.html

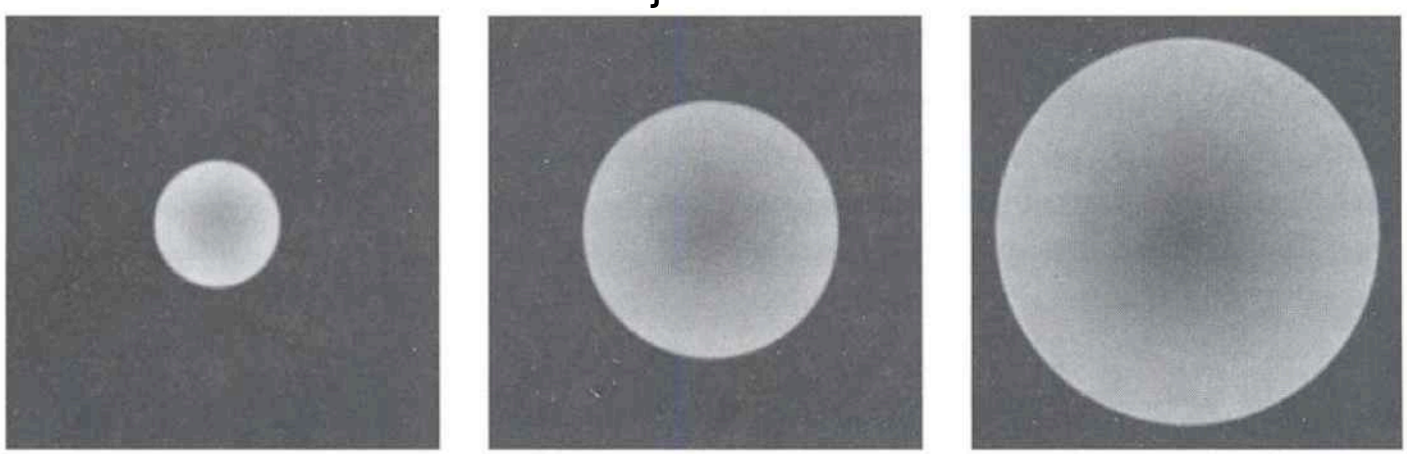
Attenuation artifact

Energy dependance



^{99m}Tc | ^{131}I | ^{201}Tl | No attenuation

Object size

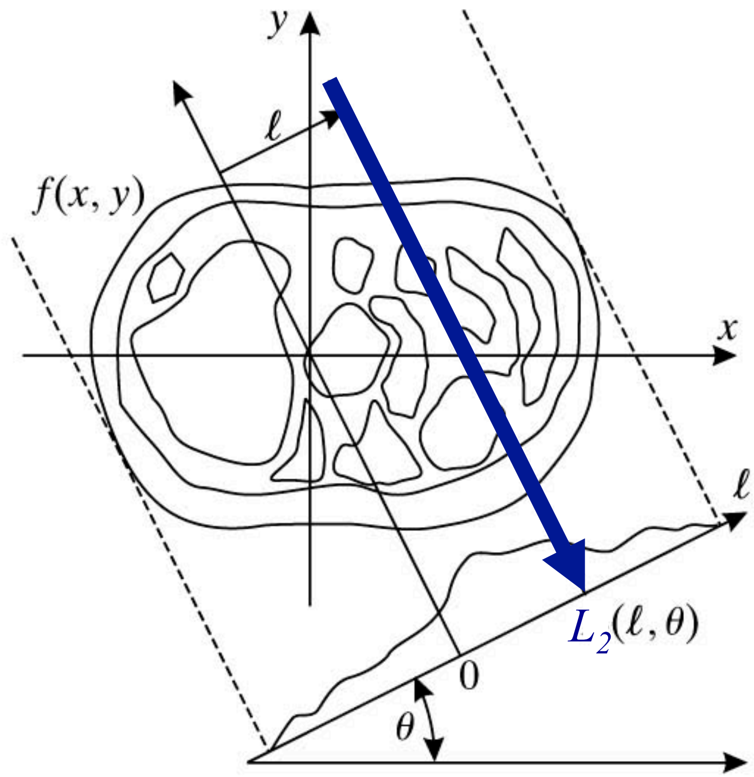


10 cm | 20 cm | 30 cm

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

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)

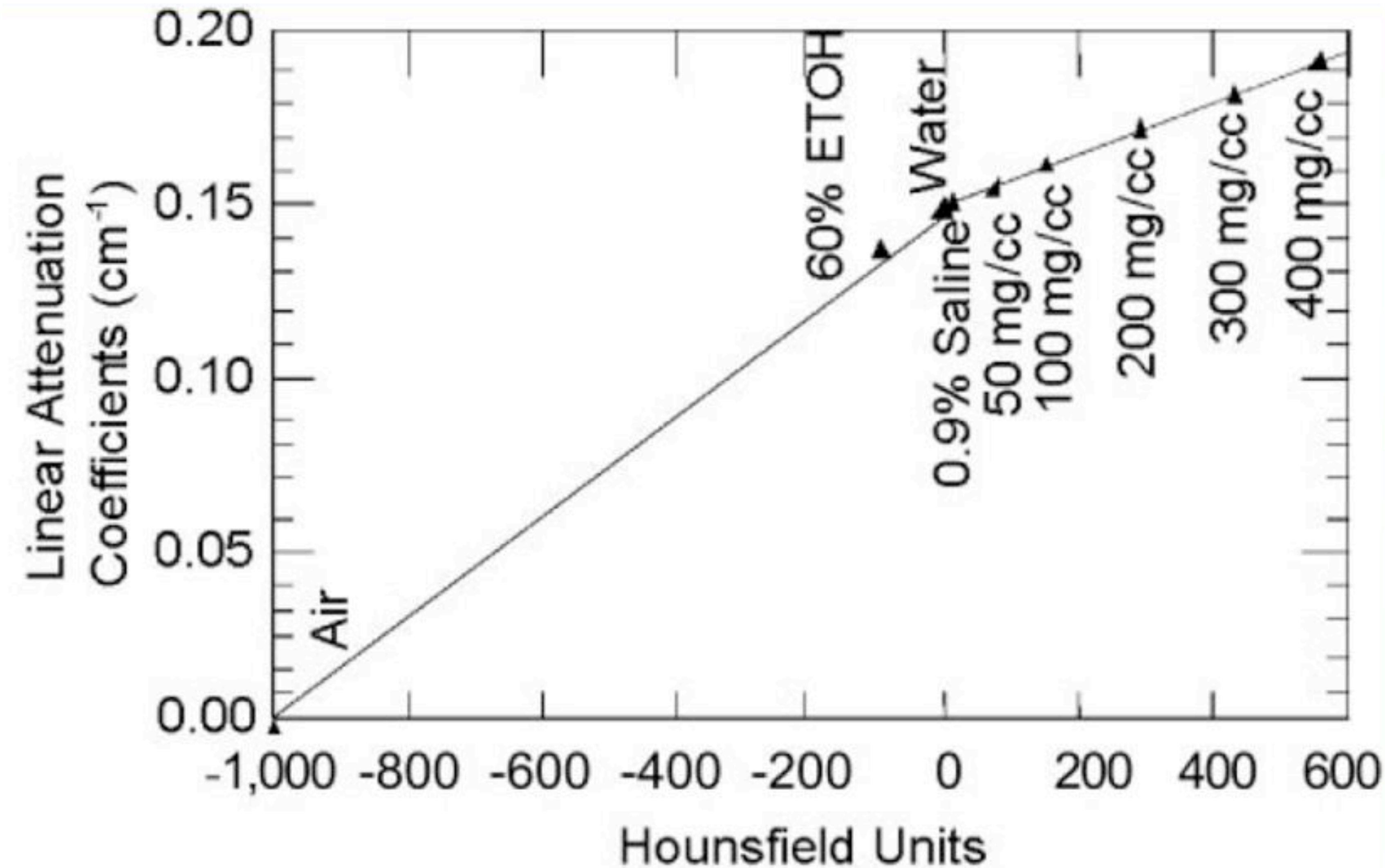
Methods of Attenuation Correction

- 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

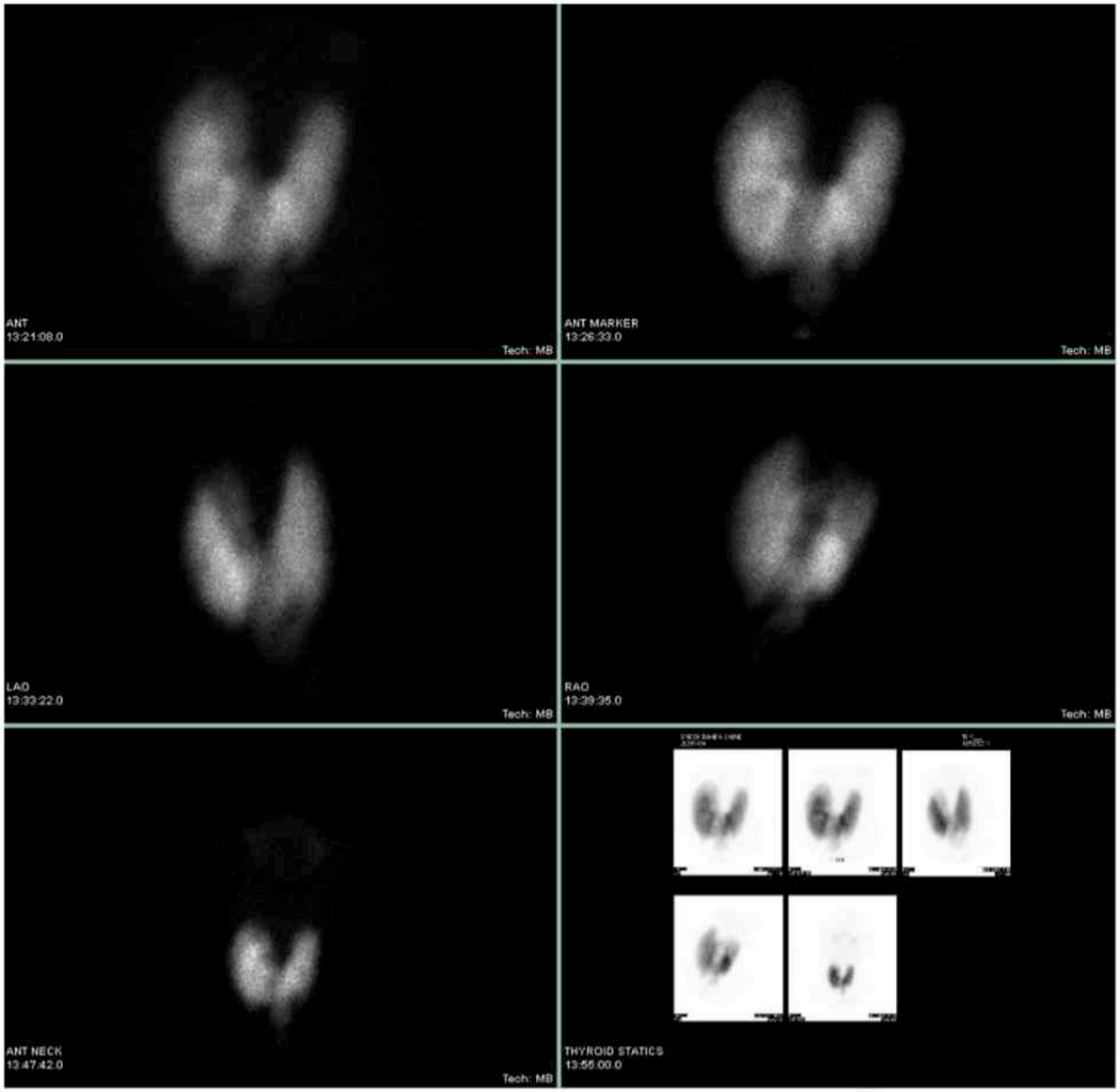
- X-ray CT
  - Co-located and co-registered
  - Lower energy than transmission delivers lower dose
  - Conversion from CT Hounsfield units to equivalent linear attenuation coefficient

Converting Hounsfield Units to attenuation coefficients



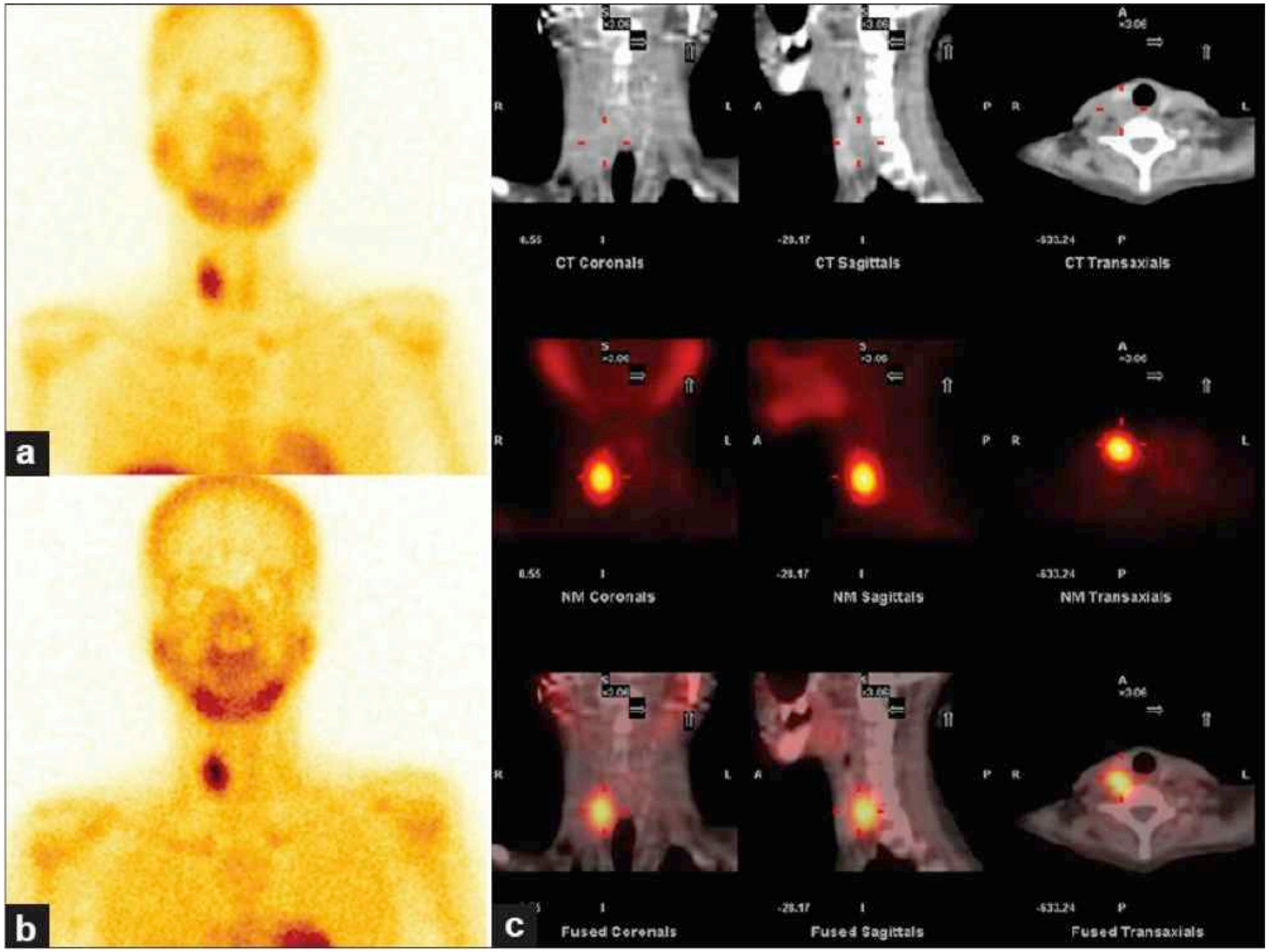
Examples of SPECT imaging

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

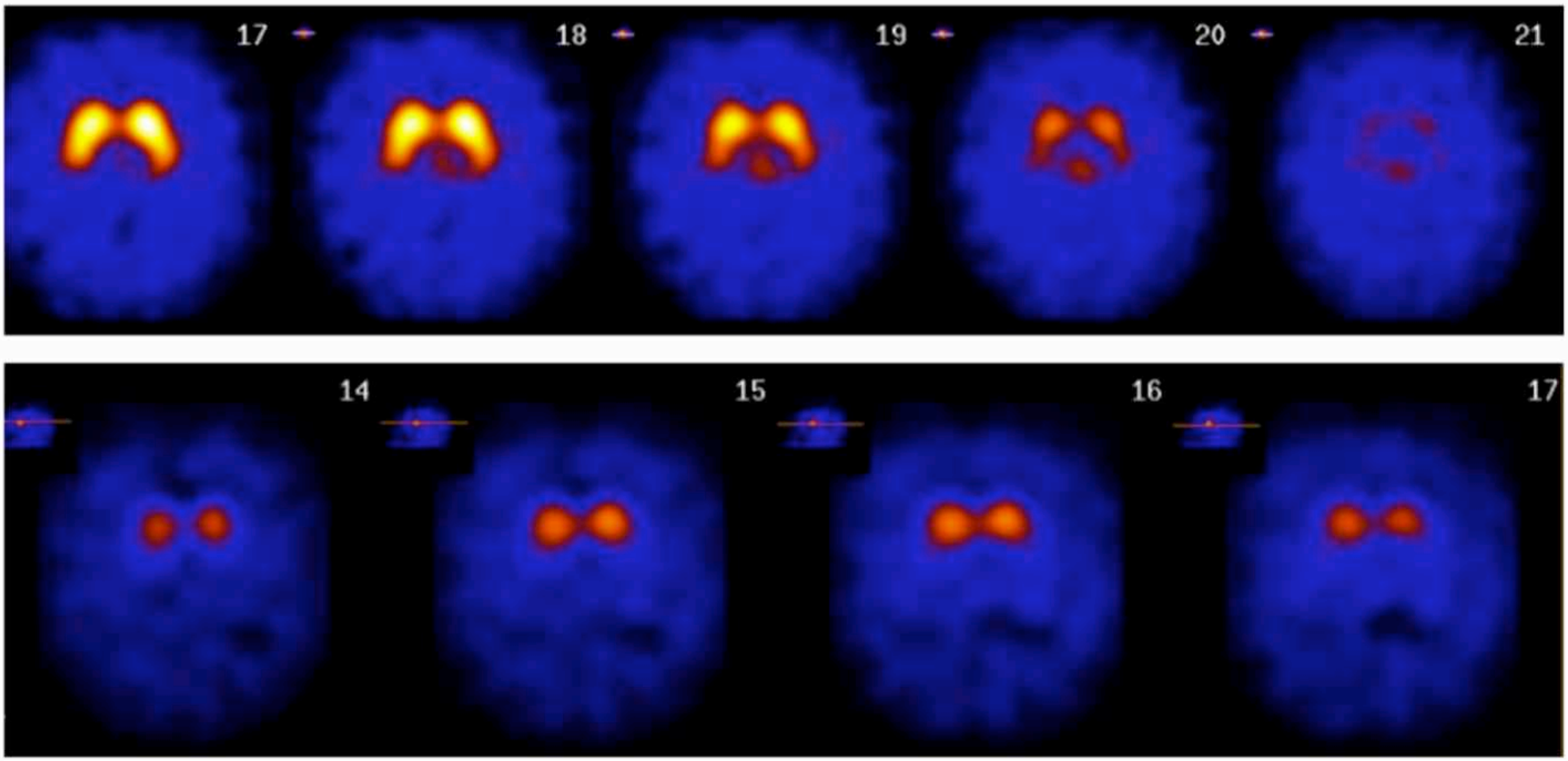
(a), (b) early and delayed Tc99m-sestamibi scintigraphs showing uptake near thyroid

(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.

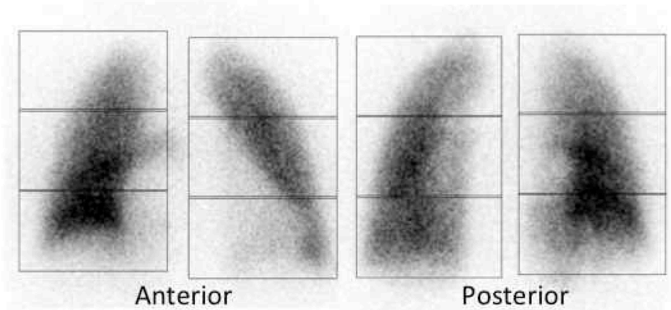
Brain Scan



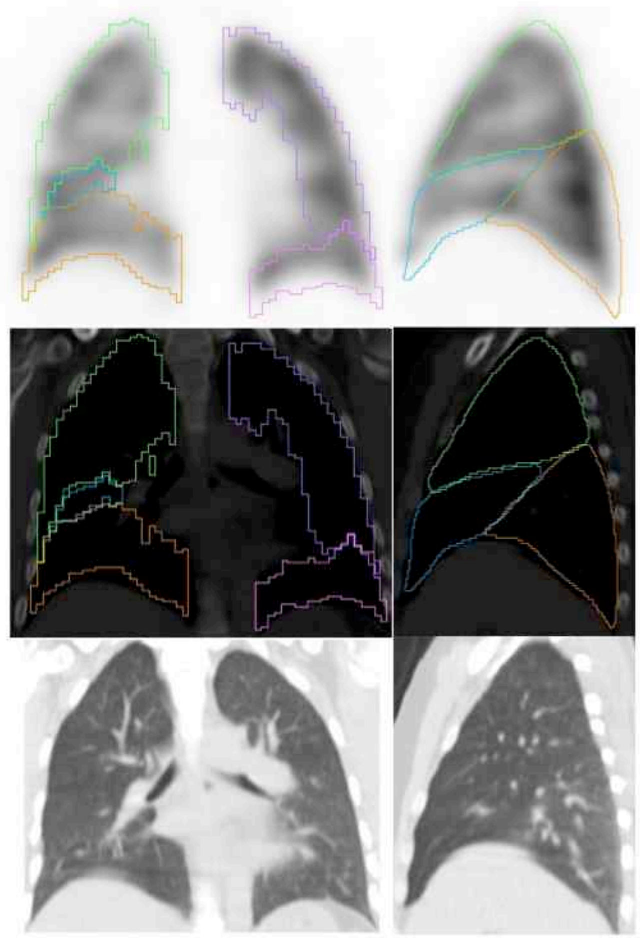
In this example, an imaging agent called DaTscan is used to differentiate between Parkinsonian syndromes (PS) and essential tremor. The "comma" image is for a patient without PS (upper panel), while the abnormal "period" scan is for a patient with PS (lower panel). DaTscan is a radiopharmaceutical indicated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

SPECT/CT Lung Perfusion Imaging

Planar Perfusion



SPECT/CT Perfusion

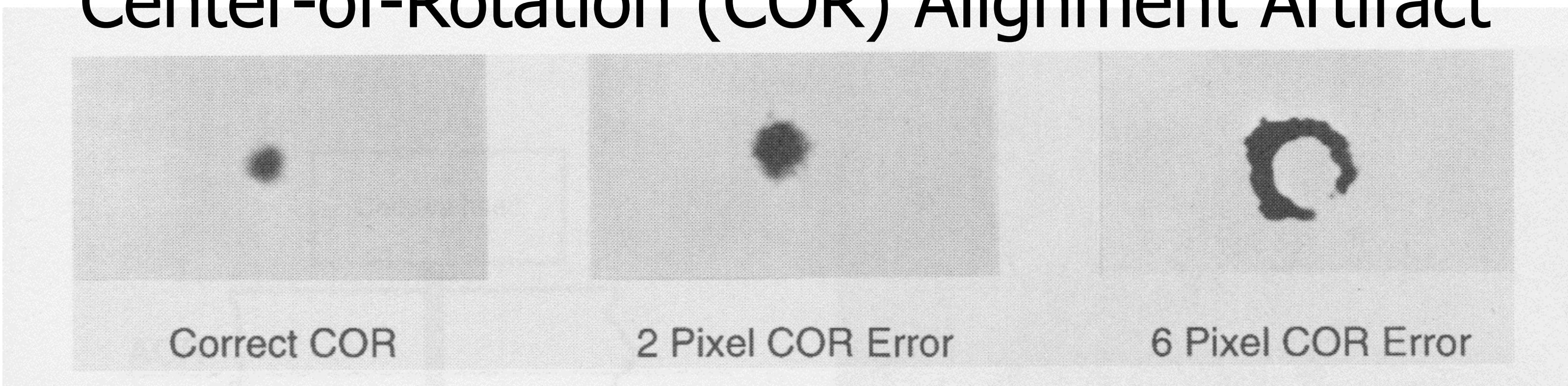


Lung zone regional perfusion estimated by geometric mean of planar counts			Lobar quantitative perfusion by attenuation-corrected SPECT/CT		
ZONES	Right	Left	**LOBES**	Right	Left
Upper 1/3	8.8%	10.4%	Upper Lobe	22.5%	24.8%
Middle 1/3	24.1%	18.9%	Middle Lobe	8.8%	-
Lower 1/3	22.4%	15.0 %	Lower Lobe	23.1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

QA/QC Tests for SPECT

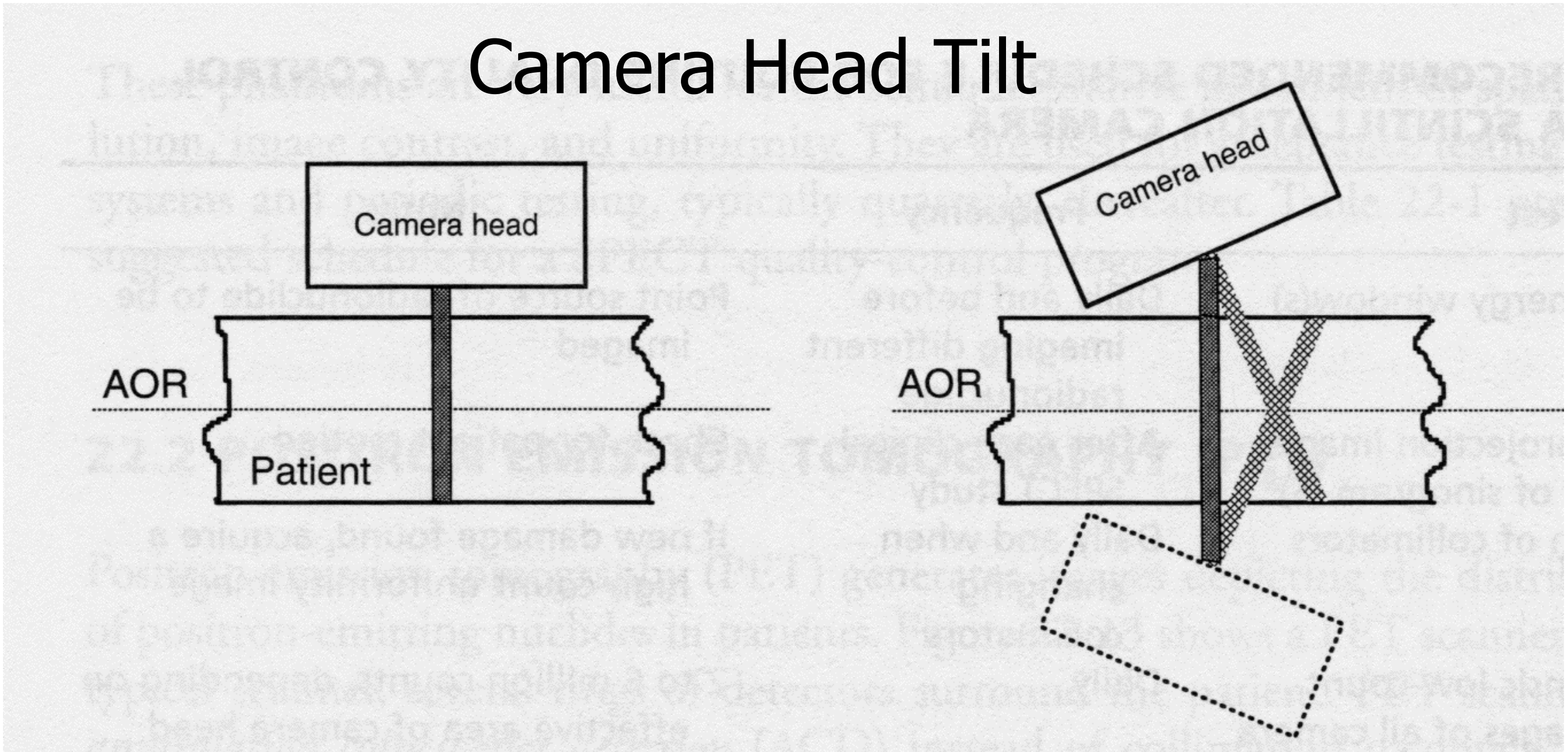
Center-of-Rotation (COR) Alignment Artifact



Small COR misalignment will blur image, while larger misalignment will cause donut shape artifacts.

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

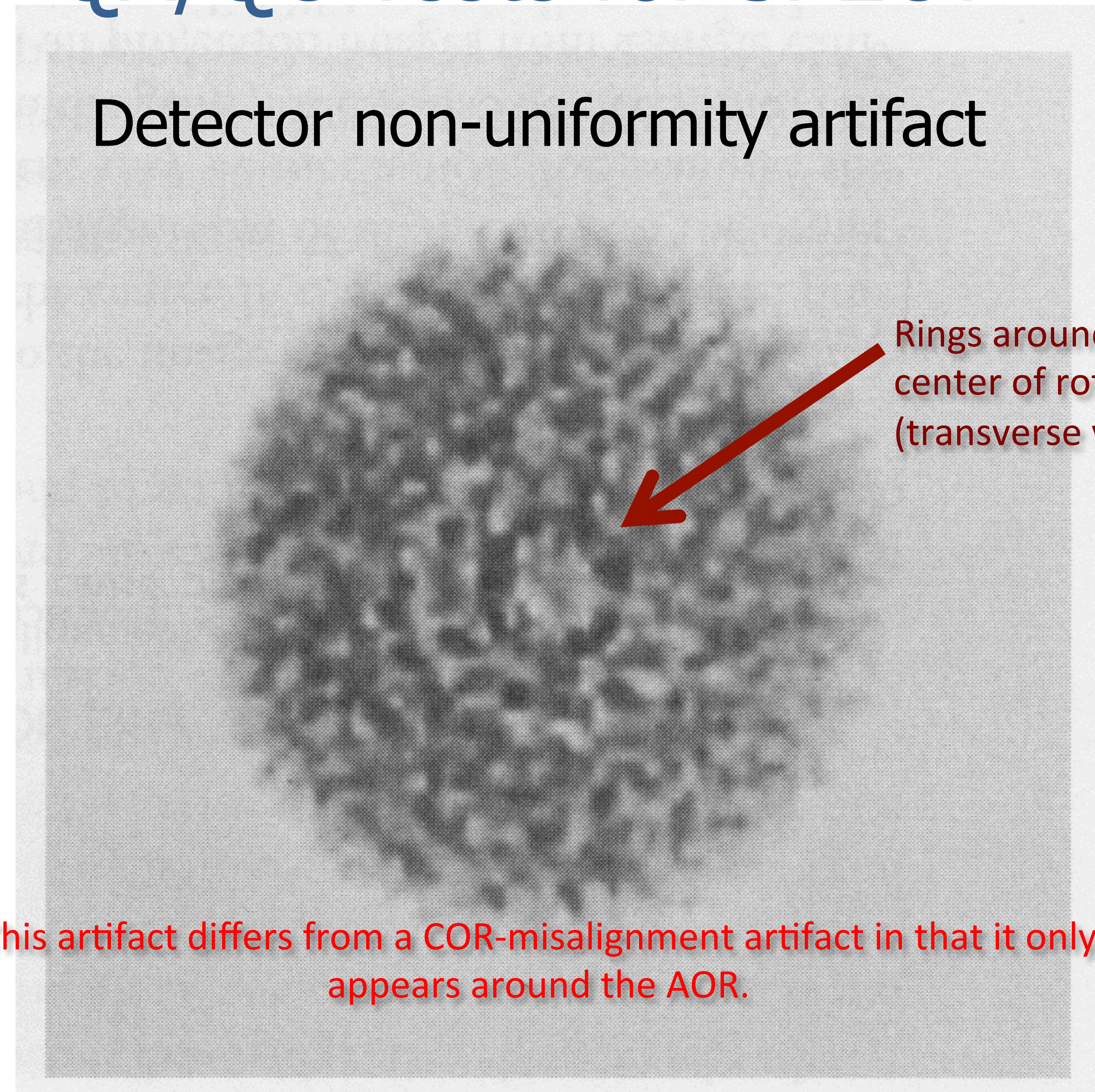
QA/QC Tests for SPECT



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

QA/QC Tests for SPECT

Detector non-uniformity artifact



This artifact differs from a COR-misalignment artifact in that it only appears around the AOR.

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

Response to Lecture 8 Questions

How do you know if it is actually properly adjusted? It showed two dots versus four dots, but how would you know if there was only suppose to be two dots and not one dot say?

If this question is in regards to multi-energy spatial registration, the answer is because we are using a known calibration source.

Response to Lecture 8 Questions

Response to Lecture 8 Questions

What determines whether Rayleigh scattering happens vs Compton scattering? In the lecture we learned in the former the entire atom is interacted with, but I am curious what this looks like on the quantum mechanics level.

The state of atomic and sub-atomic particles is described by quantum mechanics, which is a stochastic representation. The interaction of a photon with these particles is therefore also stochastic. The probability of Rayleigh, Compton, Photoelectric, or other interaction mechanisms depend on the energy dependent properties of these particles.

Compton interaction can occur with an isolated electron. The fact that the electron is bound is incidental; the incident photon is deflected and part of its energy is lost to the electron in a way that conserves momentum and energy. On the other hand, Rayleigh scatter results in electric polarization of an atom or molecule. The oscillating electric field of the photon causes charges within the particle to move at the same frequency. The particle therefore becomes a small radiating dipole whose radiation we see as scattered light. For this reason, Rayleigh scatter is more likely to occur when the photon wavelength is greater than the particle size.

Response to Lecture 8 Questions

Is there ever a time where pile-up would lead to someone misinterpreting an image and then to possibly misdiagnose a patient?

Yes. Any time the extent of an active region is confufuddled by artifact, then misdiagnosis is possible. For example, the pile-up artifact could look like connectivity of an activated lymphatic network. Alternatively, a pile-up artifact could obscure the shape and extent of an active cancerous volume, leading to poor treatment planning.

Response to Lecture 8 Questions

How does thickness of the scintillator crystal affect the spatial resolution, sensitivity and image contrast?

Scintillator thickness affects the light spread in a detector, which in turn affects the spatial sensitivity of the detector response. Thickness can also affect our depth uncertainty in the interaction position, which can lead to parallax in back-projection step of the image reconstruction. Detection efficiency is directly related to detector thickness and is expressed by an exponential relationship described as the Bear-Lambert Law. Image contrast may also be lowered for a thicker detector if we do not have an ability to reject scatter events within the detector.

Response to Lecture 8 Questions

On slide 50/87 for Lecture 8, (spatial resolution), Ri and Rc add in quadrature. Could you explain what Ri and Rc are?

Resolution of a gamma camera depends on the intrinsic (i) positioning properties of the detector (how much blur is observed for fixed interaction position) and the collimator (c) blur of the actual interaction position.

Response to Lecture 8 Questions

On the slides for the NaI crystal, there is a light guide included with it, what is the light guide?

The term "lightguide" in this context is an unfortunate misnomer. A lightguide in a gamma camera (usually a silicate glass layer) is used to light to spread for interactions occurring close to the photosensors. Allowing the light to spread to multiple photodetectors in this camera geometry allows for event positioning to a finer scale than the photodetector spacing.

Response to Lecture 8 Questions

On slide 17, what does the formula "CS *symbol* rho" mean? The slide only says the probability of Compton scattering is proportional to material density, but it does not say anything more.

On slide 17 in Lecture 8, “r” and “r” are both density. This dichotomy is due to a formatting error.

Response to Lecture 8 Questions

Are there situations in SPECT imaging where the gamma camera detector is the limiting factor on spatial resolution, or does the noise from random events and our ability to localize the radiotracer tend to outweigh this?

For a pinhole or multi-pinhole SPECT camera, detector resolution can have a significant effect on image resolution. However, for a parallel-hole collimator, it is usually collimator blur that limits image resolution. Low count rates may affect image noise properties in either case (more so for single-pin hole imaging) and therefore image resolution. Exact limitations will generally depend on the object and imaging task.

Response to Lecture 8 Questions

In PET imaging, an emitted positron interacts with an electron to produce a pair of two photons traveling in opposite directions which then can be used to give us information about the patient. Since a positron and electron can annihilate to produce two photons, can the two photons annihilate to produce a positron and electron pair (time-reversal symmetry) as well? Has this ever been observed in nature? The lecture only states the possibility of photon + nucleus and photon + electron to produce a pair.

The time-reversed process that you refer to is known as pair production. It can happen at photon energies above 1.022 MeV, but it doesn't have a significant cross-section until about 2 MeV. Pair production can also happen as a random fluctuation in the electronic or nuclear fields around an atom. These so called "vacuum fluctuations" are responsible for an experimentally observed phenomena known as the "Lamb Shift" which is a fine-scale splitting of electron energy levels in an atom.

Response to Lecture 8 Questions

On Slide 28, would you explain how mu = 0.15 /cm was determined?

Attenuation coefficient data for different elements are available from NIST:

http://www.nist.gov/pml/data/xraycoef/index.cfm

http://physics.nist.gov/PhysRefData/Xcom/html/xcom1.html

The value of 0.15/cm is the product of the mass-attenuation coefficient for the water molecule and the density of water.

Response to Lecture 8 Questions

How does a pinhole collimator magnify an X-ray image?

Magnification of a pinhole image is due to the relative distances of source and detector from the pinhole. Photons traveling lines through the pinhole produce an inverted image of the source distribution. This projected image through the pinhole expands as the detector is moved back. Unit magnification occurs when the detector and source are equidistant from the pinhole.

Response to Lecture 8 Questions

What features of an imaging device affect or define its energy resolution?

Energy resolution of a detector depends largely on the number of signal carriers (i.e. scintillation photons or electron-hole pairs). Some scintillators can have a non-proportional response due to stochastic of the scintillation process. This non-proportional response can lead to increased signal variance. For a scintillation camera, geometry and photodetection efficiency can also play a significant part in the light collection efficiency and therefore the energy resolution. Dark current and saturation effects of the photodetectors can also play a part. Electronic noise of the readout electronics is not usually significant if well designed.

Response to Lecture 8 Questions

Why is there a large scintillator over all PMTs in gamma cameras? Why not have them separated?

In older gamma camera designs, a monolithic scintillator was used because the photomultipliers were large and there needed to be enough light spread to allow for event positioning to a scale smaller than the phototube. Newer camera designs have used discrete crystal arrays coupled to smaller photodetectors. Some designs have used one-to-one coupling of crystal elements and photodetectors. Others have some light sharing to allow for crystal element identification in a similar manner to the older gamma cameras.

Response to Lecture 8 Questions

Are there any spectroscopic gamma cameras?

Yes. Multiple energies can be resolved with gamma cameras. Energy resolution of scintillation cameras are in the range of 7-15% depending on the scintillator and photodetector used. Semiconductor gamma ray detectors such as CZT, Silicon, or Germanium provide much better energy resolution (2% or better) but their detection efficiency suffers due to the low average atomic number.

Response to Lecture 8 Questions

Regarding the crystal in a scintillation camera, what is (are) the characterization(s) that determine the optimum thickness of crystal?

Incident photon energy is the primary consideration. Higher energy photons will require thicker detectors for sufficient detection efficiency. High detection efficiency helps to reduce patient dose needed to produce a useable image. However, a thick detector can degrade intrinsic detector resolution, can cause parallax due to lack of depth resolution, and can also cost more. Therefore, choice of scintillator thickness will be a tradeoff of detection efficiency and detector resolution that will need to be optimized based on the task (considering image resolution requirements, count rate, field of view size, and depth-of-interaction resolvability).

Response to Lecture 8 Questions

Are scintillation detectors more common than solid state detectors because of cost alone? I would imagine the resolution you could get from solid state detectors would be significantly better. If costs where to decrease, would there be (or is there already) a shift to solid state detectors?

Cost is a significant factor. However detection efficiency is a significant issue due to their lower average atomic number.

Response to Lecture 8 Questions

Question on the Sample Spectroscopy System slide:

Why does the circuit consist of a pre-amplifier and an amplifier?

A preamp is a high-input-impedance amp with highly linear response. It is used to buffer the input and amplify the signal to levels that is well above the noise floor of downstream electronics. The pre-amp is often co-located with the detector to minimize corruption of our signal when it is at its weakest amplitude. A second stage amplifier is used to shape the impulse characteristics to a form that is expected by the data acquisition electronics (DAQ) or multi-channel analyzer (MCA). This second stage does not have to be collocated with the detector, which may alleviates space constraints of the imaging or spectral device.

Response to Lecture 8 Questions

Since gamma cameras can allow for simultaneous imaging of radiotracers of different energies. Is there a time when the imaging would be required to be done simultaneously opposed to having the radiotracers injected simultaneously and then scheduling the scans to get the scans done as soon as possible?

When image registration is critical and/or relative timing the metabolism of these tracers is important, then simultaneous imaging of spectrally separable tracers is very important.

Response to Lecture 8 Questions

This question references slide 56. How can better than 4 mm resolution be obtained when the PMT's are much larger than 4 mm.

A gamma camera typically consists of a monolithic scintillator readout by an array of photomultiplier tubes (PMT). The light from a scintillation pulse distributes itself to more than one PMT. Though the amount of light reaching each tube is stochastic, the average distribution of light is a function of position. The average responses of the PMT array are typically sensitive to position offsets smaller than the phototube spacing. For instance, if we compute interaction position from the energy-weighted centroid, then a small shift in the signal distribution can offset our position estimate by less than the PMT spacing.

Response to Lecture 8 Questions

Are collimators really built to be essentially a grid of with boxy rectangular walls as shown in the slides, or is that a simple representation of the collimators used? It seems like if they had a curve that transitioned into a slim channel that waves could be guided into the collimator and thus improve the gain of the detector (probably at the cost of some contrast...)

Cylindrical wall are often used for parallel hole collimators. The bore-hole shape of other collimators (diverging, converging and pinhole) are all more complex. Converging and diverging bore holes tend to be conical and can be tilted. Pinholes tend to be hourglass shaped, tend to have keel thickness, and can be tilted.

Response to Lecture 8 Questions

Slide 15 states that photons are preferentially absorbed by more tightly bound (inner shell) electrons. I understand why there is an absorption peak at the binding energy of an electron shell, but I don't see why the inner shells would preferentially absorb photons versus any shell. Can you elaborate on this assertion?

Momentum cannot be conserved if an isolated electron were to absorb all the energy of a photon. Momentum transfer to the nucleus that binds an electron makes this interaction possible. Compared with more loosely bound electrons, an inner shell electron spends more of its time in close proximity to the nucleus. Colocation of the electron and nucleus results in a higher cross-section (higher probability) for photoelectric interaction.

Response to Lecture 8 Questions

According to D75, a star-burst artifact would be due to the hexagonal arrangement of holes in the collimator. Is there a geometric arrangement that would minimize such occurrences?

Septal penetration causes image blur. This blur exhibits symmetries that reflect the collimator geometry (more blur in directions that wall thickness is thinnest). In some sense, an ideal collimator exhibit is one that balances collimator blur and photodetection efficiency. There is no static collimator design I know that exhibits isotropic blurring. However, perhaps spinning the collimator might reduce this type of artifact.

Response to Lecture 8 Questions

When using gated techniques it says that 16 to 24 frames a sequence are typically acquired Is there a threshold where more images in that sequence aren't providing any additional information?

For image-based estimate, it is best to have as many counts per frame to reduce image noise. However, to limit motion blur, the number of frames in a gated dynamic sequence will depend on the range of motion in the dynamic sequence compared to the expected motion-free image resolution.