Structure of the course

1) Introduction
2) Therapy with proton and ion beams
3) X- ray sources
4) Sources for nuclear medicine
5) Detection of photons (physics and detectors)
6) Image quality
7) Image reconstruction
8) X-ray imaging
9) **Planar scintigraphy**
10) Computed tomography
11) Emission tomography (**SPECT**, PET)
12) Magnetic Resonance Imaging
13) Multimodal systems

sources

principles / tools

objective

method

imaging modalities

Medical imaging

* Prince and Links, Medical Imaging Signals and Systems, Chap. 8,9.

2D projections

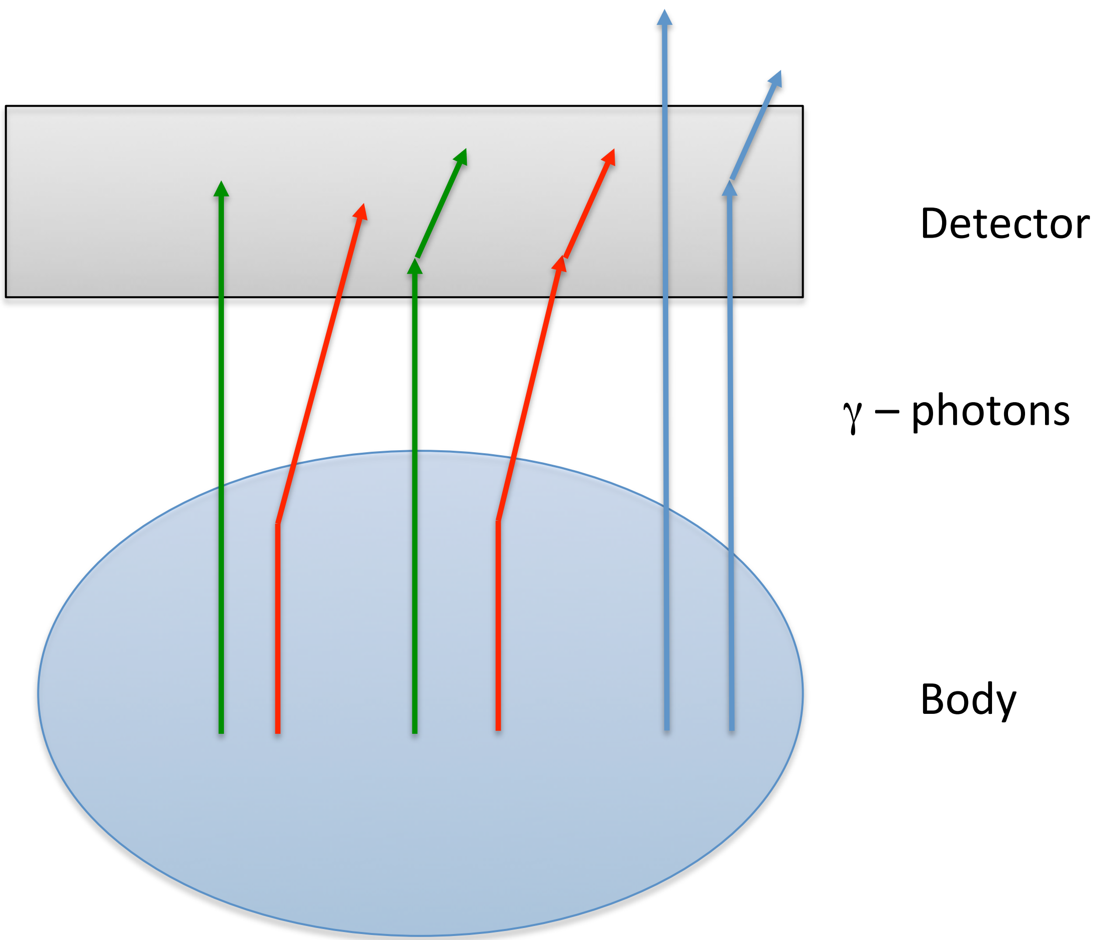
- X-ray & planar scintigraphy are two types of 2D projection imaging

Imaging principle:

Planar scintigraphy: By capturing the emitted gamma photons in one particular direction, determine the radioactivity distribution within the body

X-ray imaging: determine the attenuation coefficient to the x-ray

Recap on photon interactions



- Photo-electric effect
- Rayleigh (elastic scattering)
- Compton (inelastic scat.):

$$E_\gamma{}'_{(\max)} = \frac{E_\gamma}{1+2\varepsilon}$$

$$\varepsilon = \frac{E_\gamma}{m_e c^2}$$

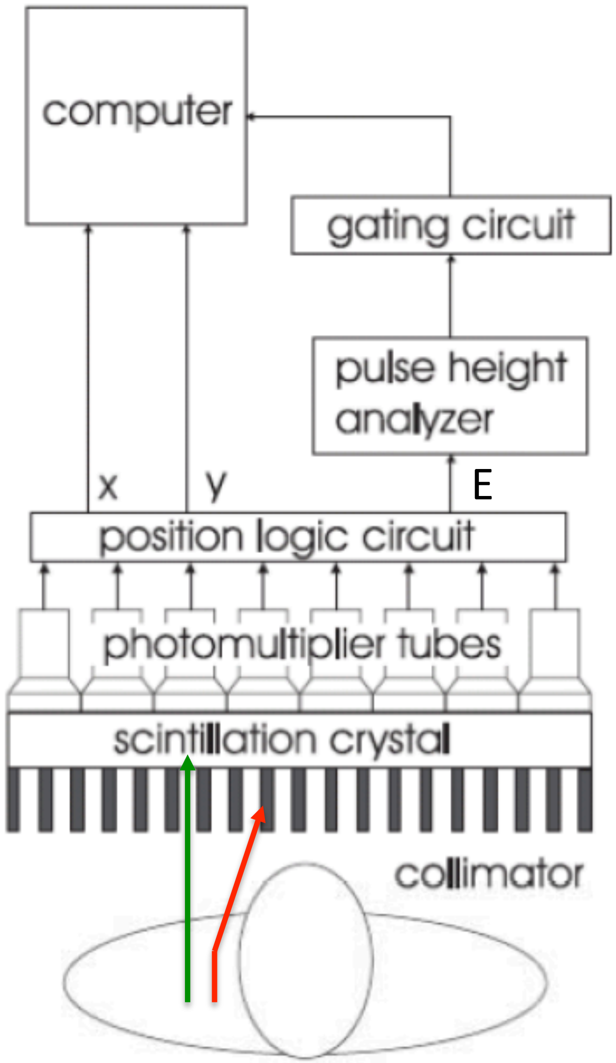
Example ^{99m}Tc:

$E_\gamma =$ 140 keV

E'_γ (max) = 90 keV

Anger scintillation camera

Hal Anger, 1950



➔ Record the event as (x,y,E)

➔ Compute the total charge (energy of gamma) and compare it to a threshold (minimum/maximum energy cut)

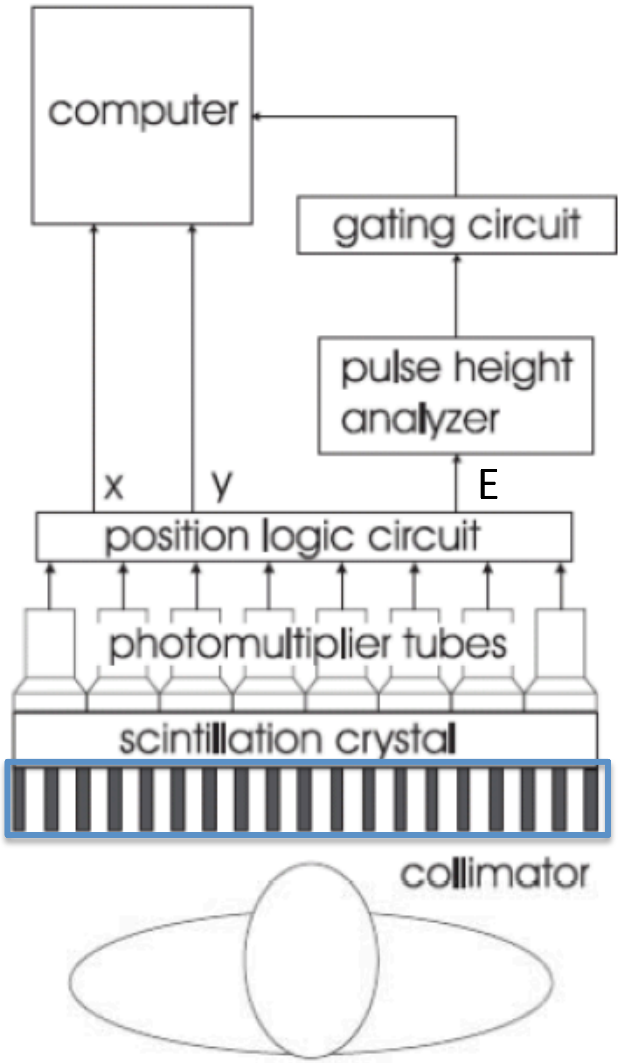
➔ Compute the location with highest activity

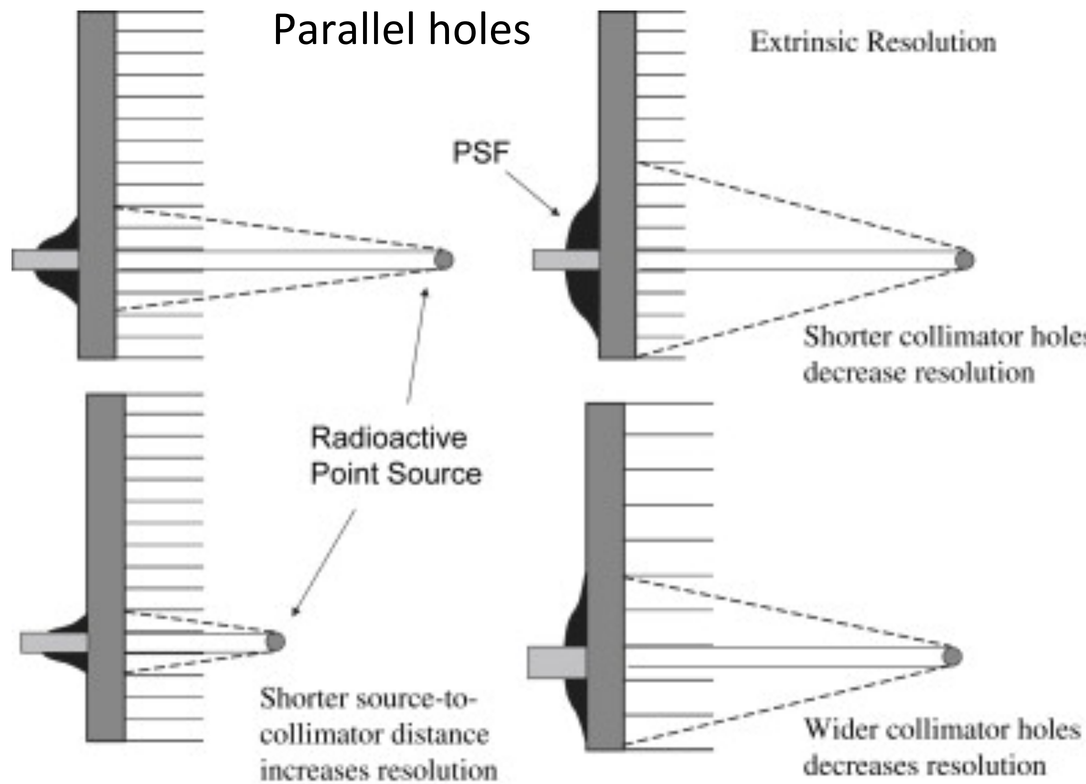
➔ Convert light into electrical charge

➔ Convert gamma photons into light

➔ Absorb scattered photons

Collimators



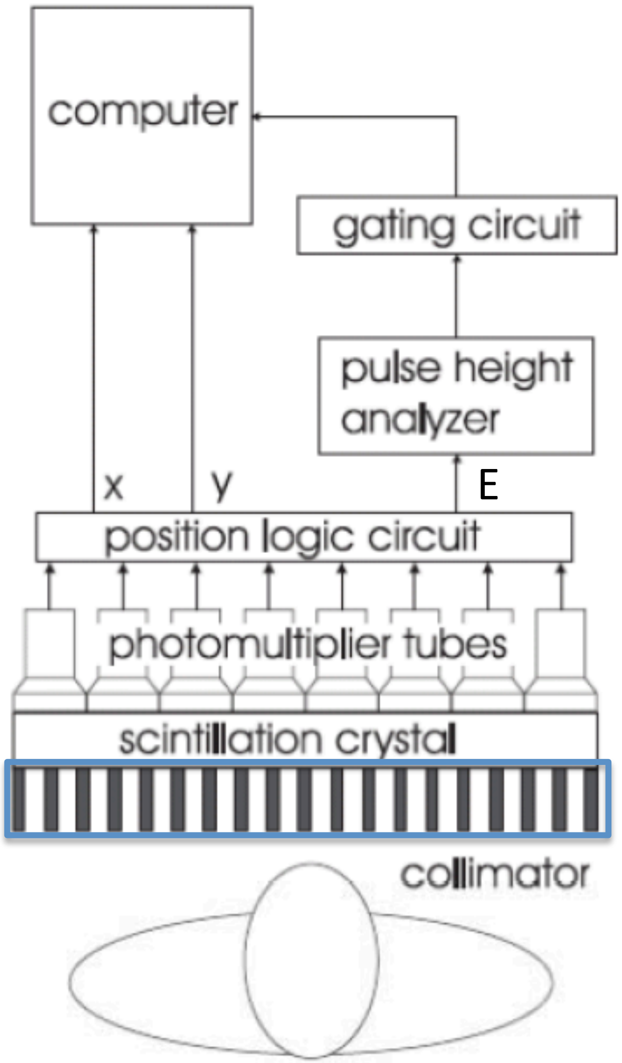


➔ Absorb scattered photons

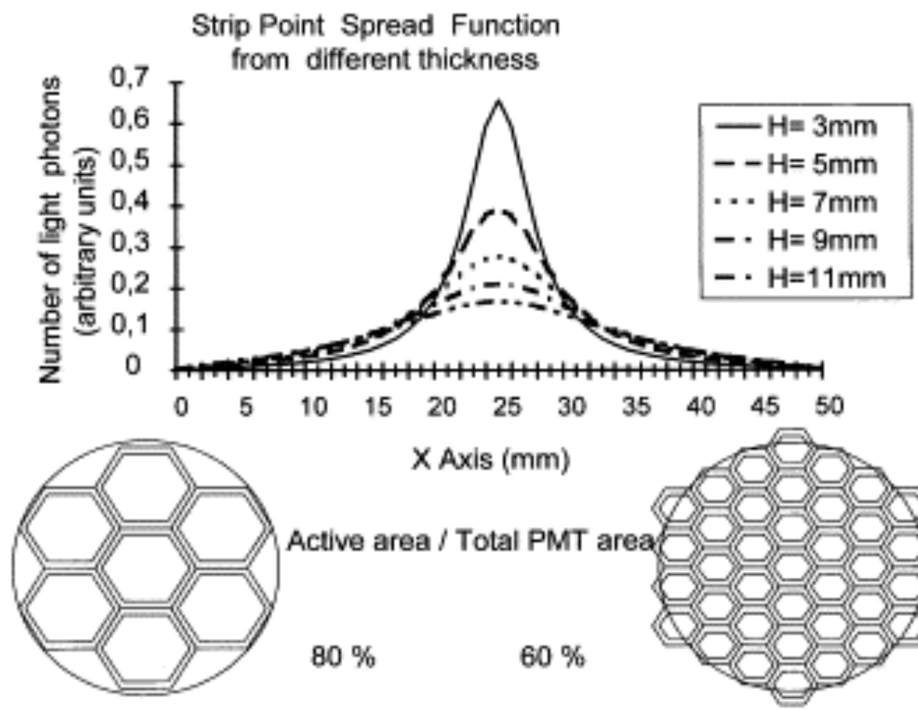
Also other type of collimators exist:

- Converging holes (magnifies)
- Diverging holes (minifies)
- Pin-hole (2-5 mm)

Collimators



Parallel holes



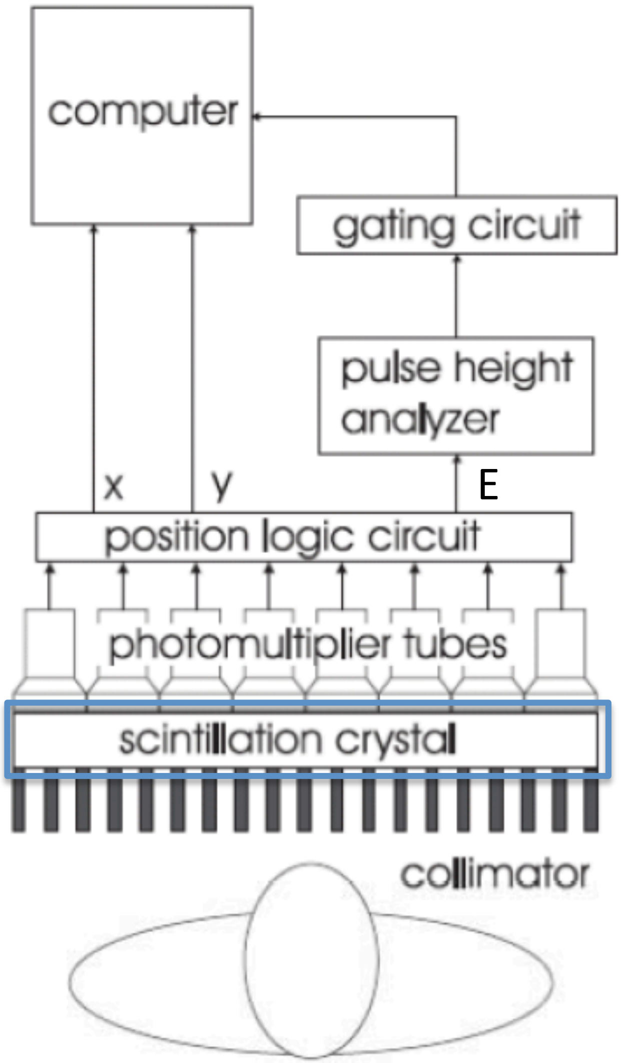
➔ Absorb scattered photons

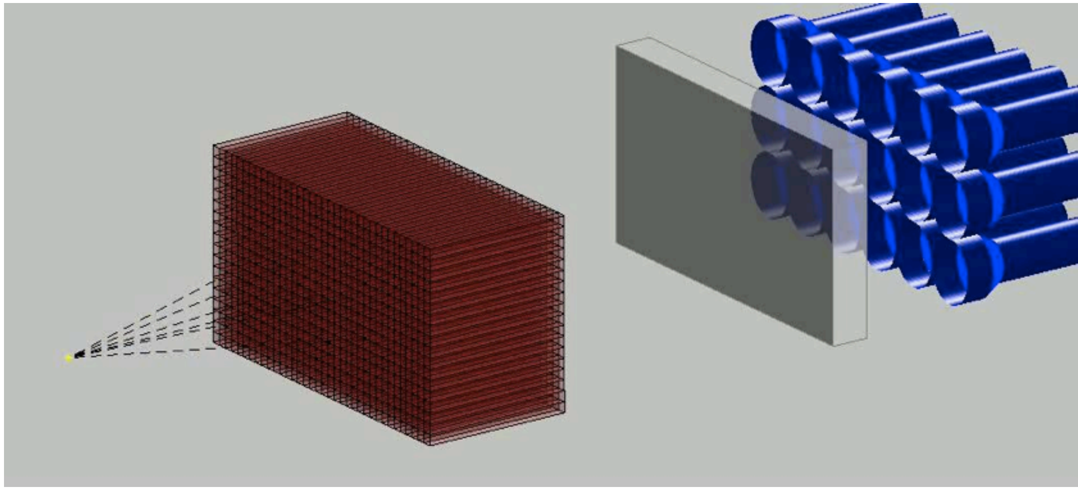
Square or hexagonal shaped holes to maximize homogeneity

Large holes = more sensitivity / less resolution

Small holes = less sensitivity / more resolution

Scintillator

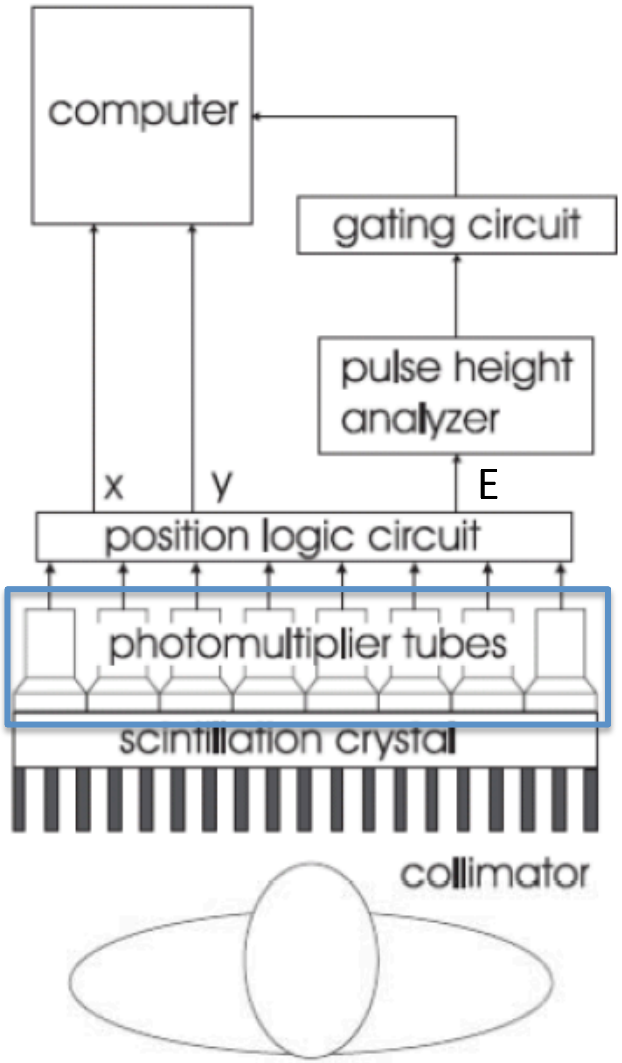


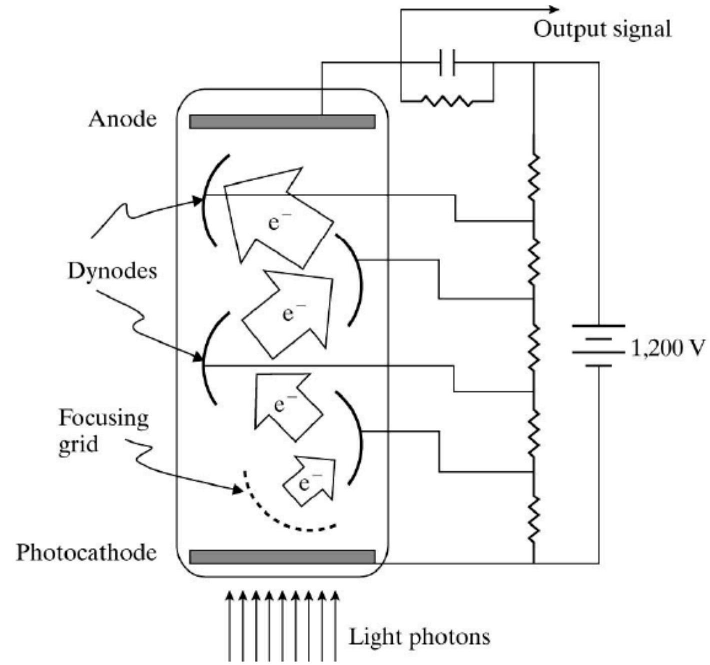


➔ Convert gamma photons into light

Single plate (30-50 cm) crystal: NaI(Ta)
Thickness:
High-energy emitters: 12.5 mm
Low-energy emitters: 6-8 mm
Produced light spreads over the whole crystal
Detected by all photo-tubes

Photomultipliers tubes

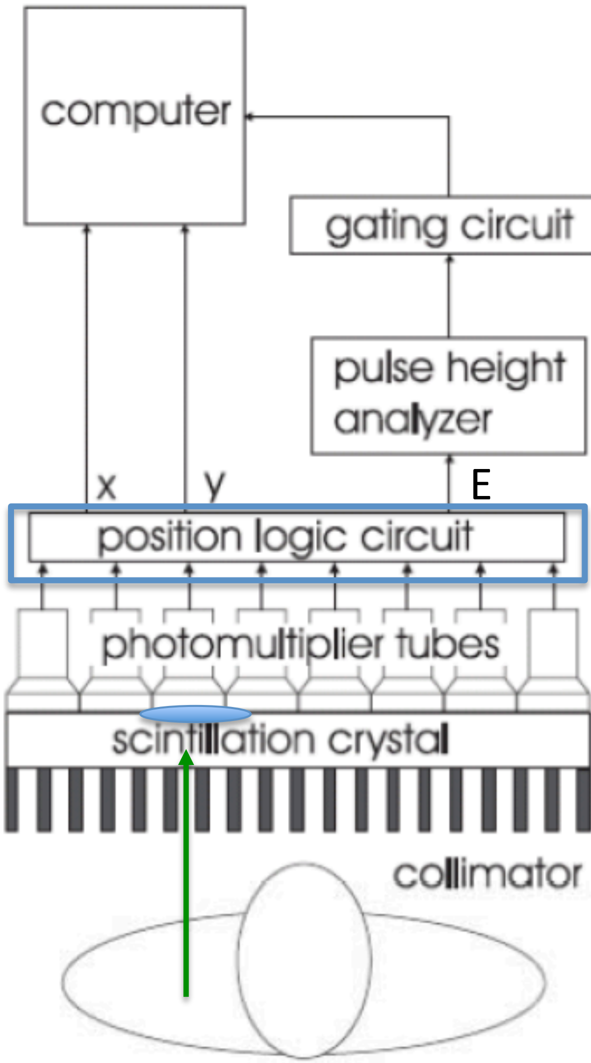


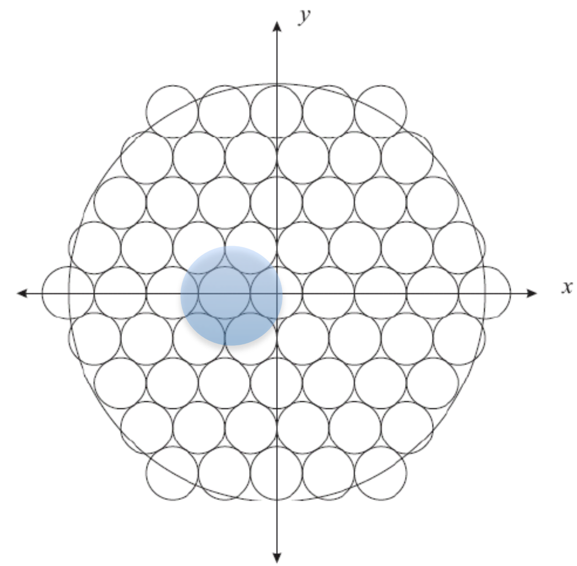


→ Convert light into electrical charge

- Each tube converts a part of light into an amplified electrical signal
- The sum of all signals (or total charge) is proportional to the energy of the gamma photon

Position logic circuit





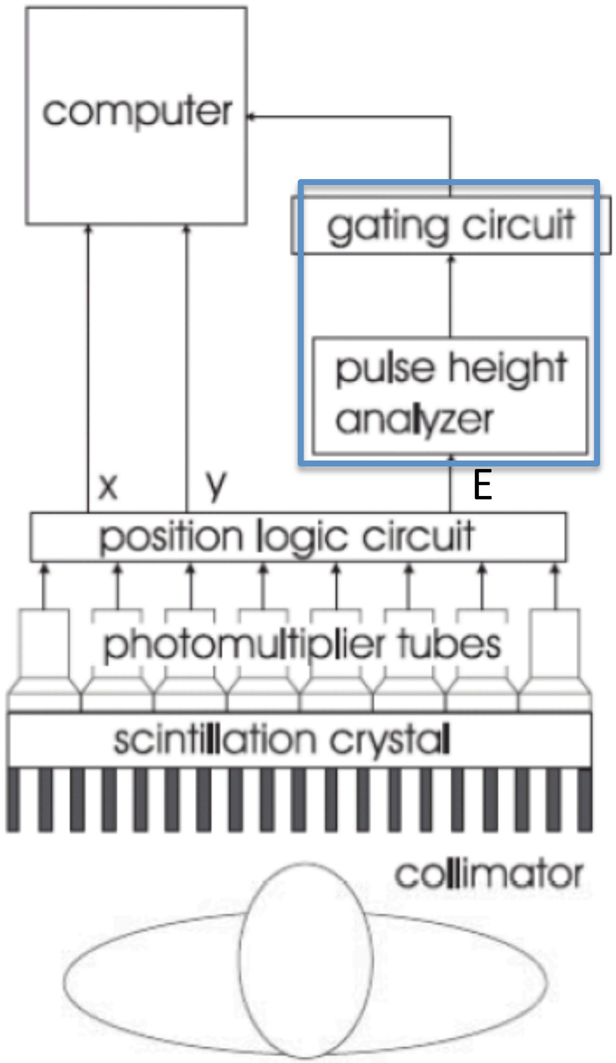
➔ Compute the location with highest activity

Tubes centers at (x_k, y_k) for $k=1,\ldots N$
Center of mass of the pulse:

$$X = \frac{1}{E}\sum_k x_k E_k$$

$$Y = \frac{1}{E}\sum_k y_k E_k$$

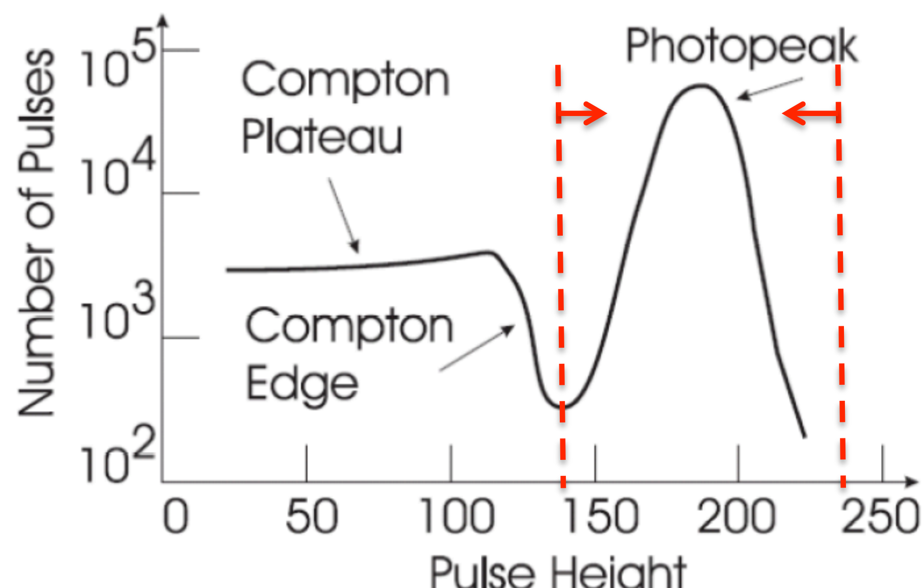
Pulse height analyzer



The sum of all photo-multipliers signals (calibrated to energy) gives the gamma energy:

$$E = \sum_k E_k$$

➔ Compute the total charge (energy of gamma) and compare it to a threshold (minimum/maximum energy cut)



Event rejection

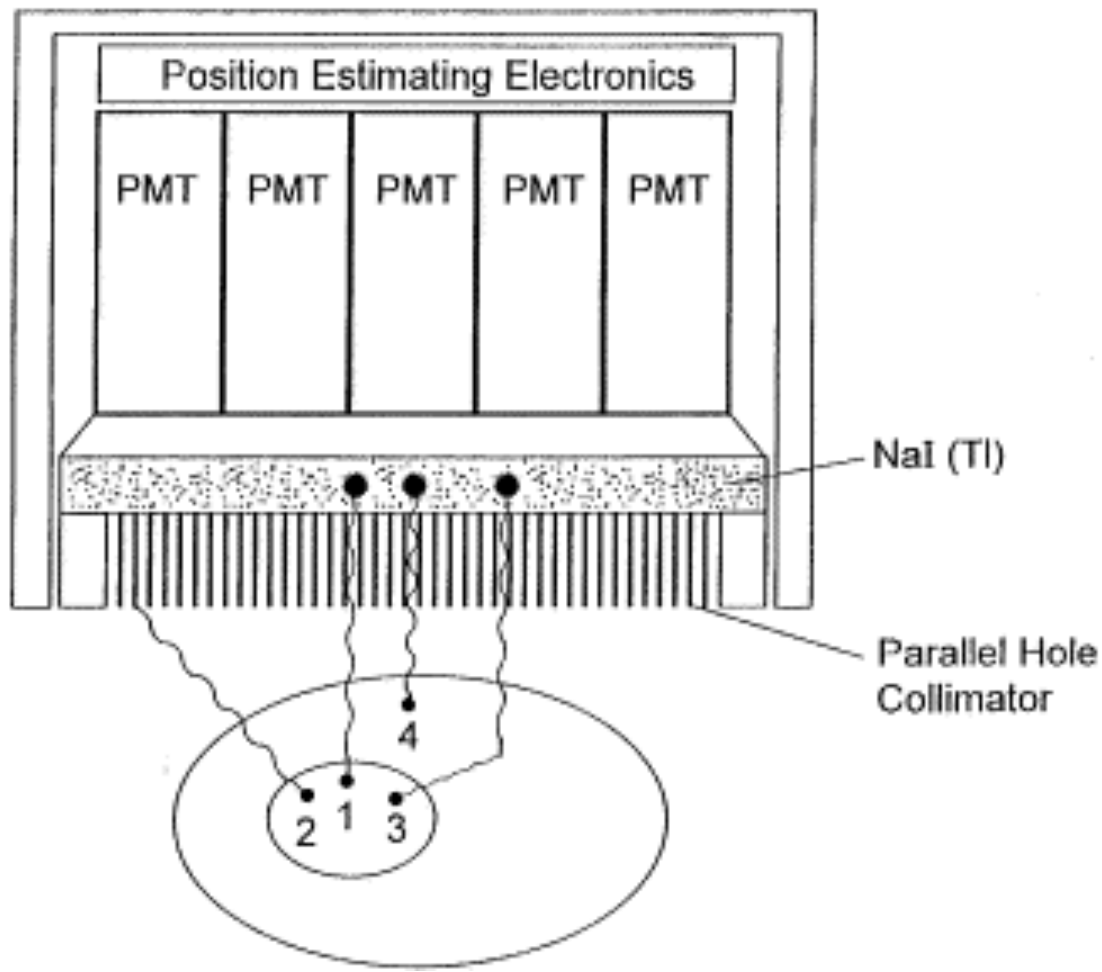
Collimator rejection:

- Define the line of response
- Reject Reyleigh scattered γ
- Reject most Compton scat. γ

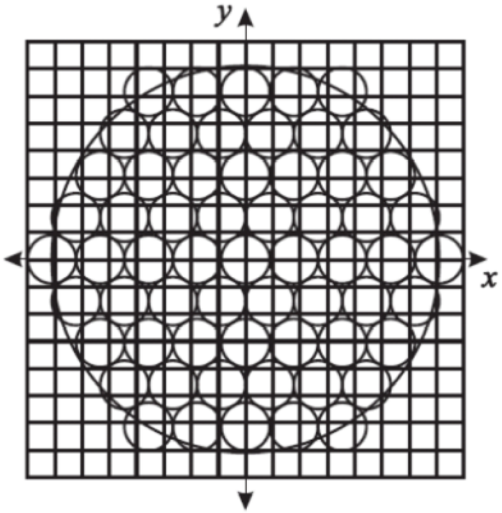
Energy rejection:

- Remaining Compton scat. γ
- Wrong energy photons
- Double photons

Event # 4 would not be rejected but does not come from the region under study



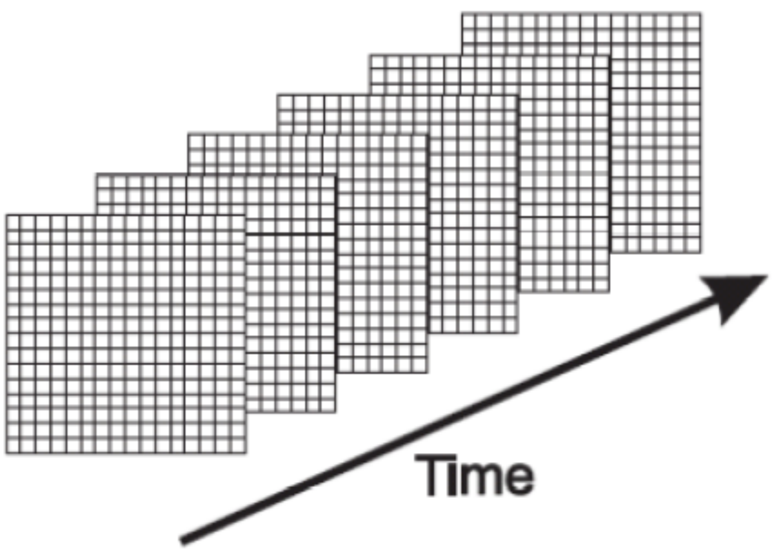
Acquisition mode



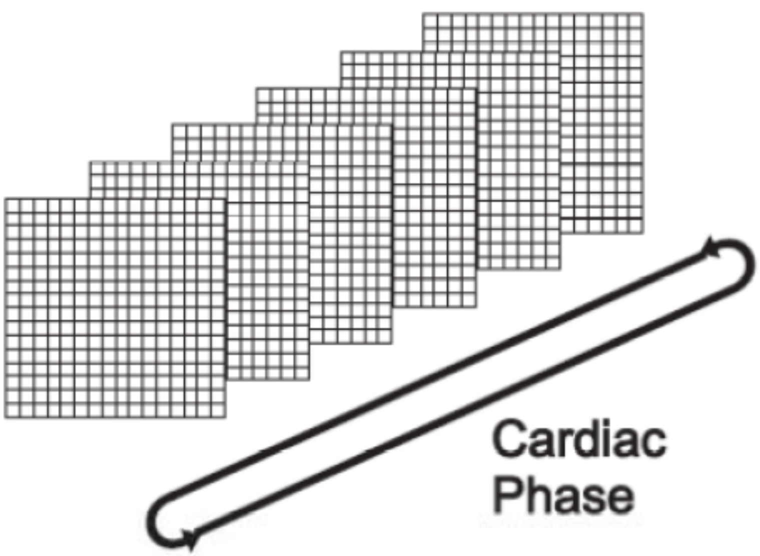
Static frame: delivers for each pixel the number of events happened in that location over the entire scan time
Matrix size: 30x30 or 50x50 cm^2 with 64x64 – 256x256 pixels

Larger area covered by multiple static frame acquisitions

Dynamic frame mode:

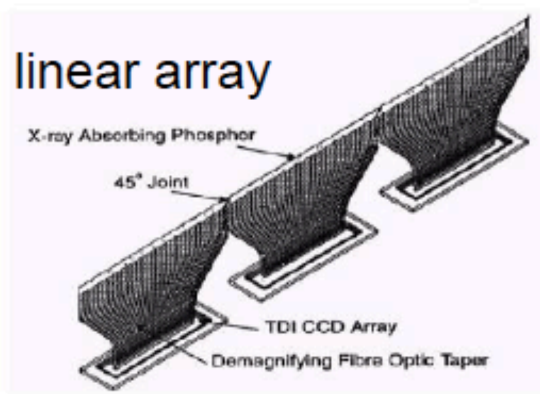


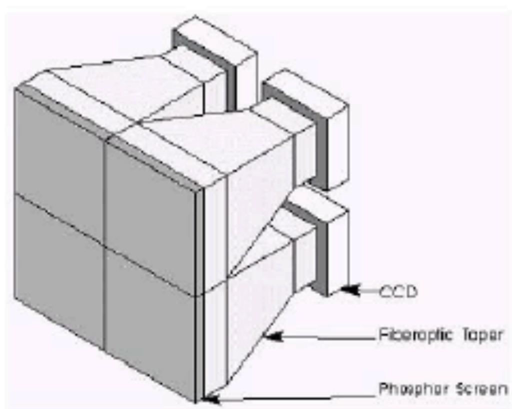
Multiple-gated acquisition:



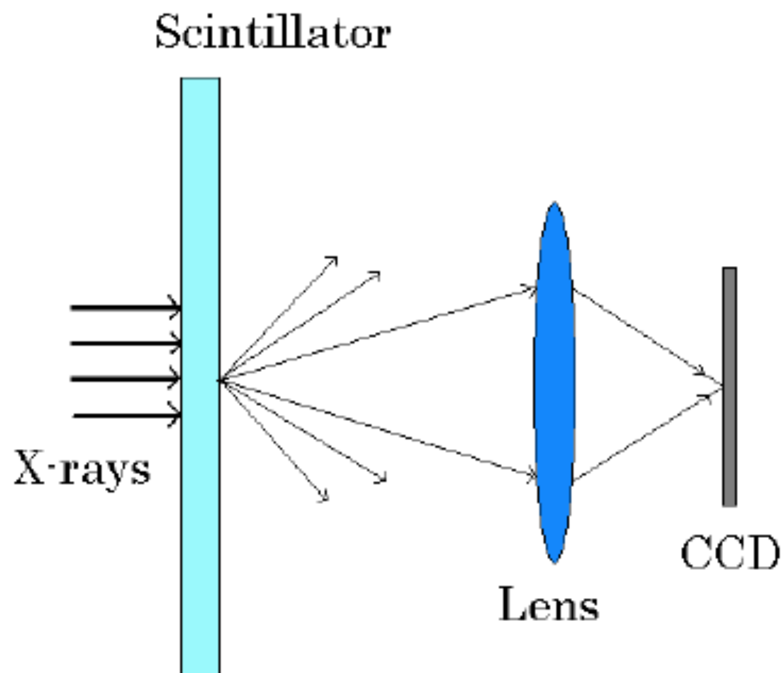
Compare to X-ray detector

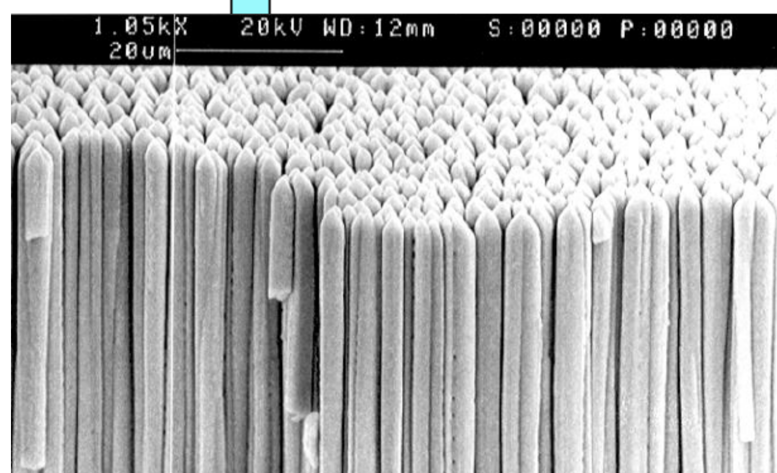
CCD linear array





CCD camera system

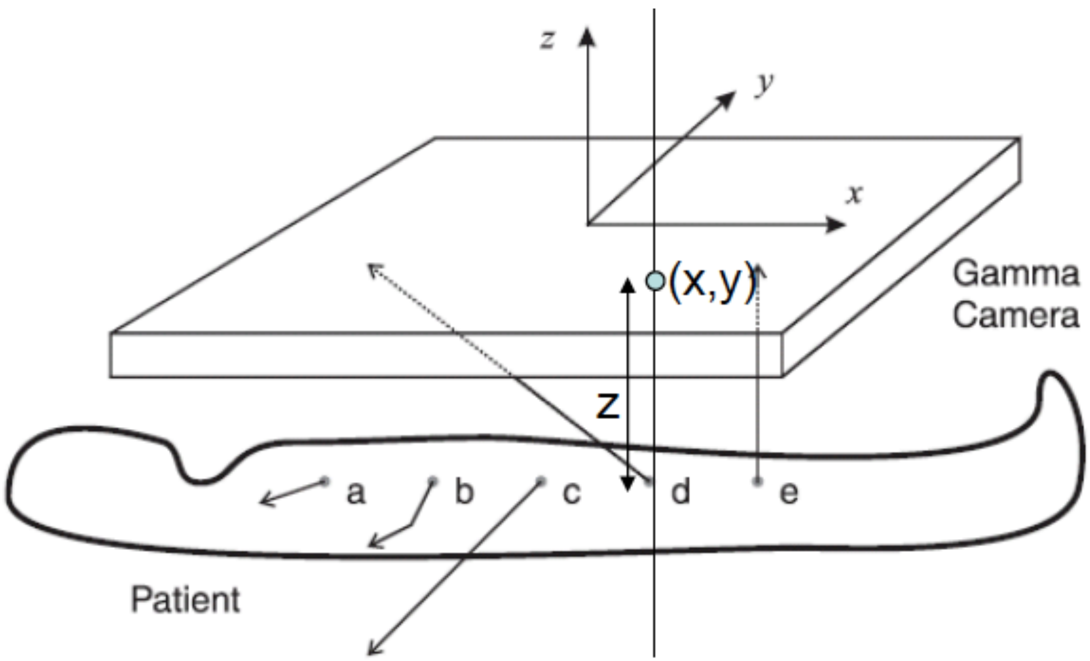




QE

Photon detection via:
- phosphor screen or
- CsI:Tl needle crystals 550 um thick ➔

Imaging Geometry and Assumption



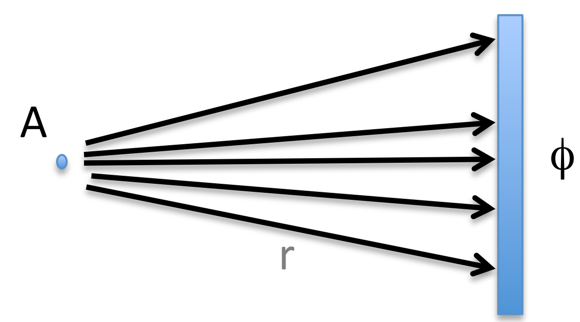
- Lines defined by (parallel) collimator holes
- Ignore Compton scattering
- Radioactivity is $A(x, y, z)$
- Monoenergetic photons, energy E

Imaging Equation

Given the activity A=A(x,y,z) in a point within the body the intensity of photons on the detector is photon fluence rate times photon energy:

$$I_d = \phi_d \cdot E = \frac{A}{4\pi r^2} \cdot E$$

This formula neglects photon attenuation



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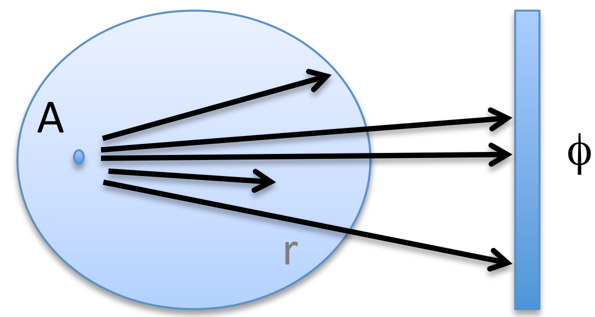
Including body attenuation the photon flux is reduced:

$$\phi_d = \frac{A}{4\pi r^2} e^{-\int_0^r \mu(s;E)ds}$$

Mono-energetic assumption
= no integral over E

Different from X-ray imaging where
r is fixed = source position



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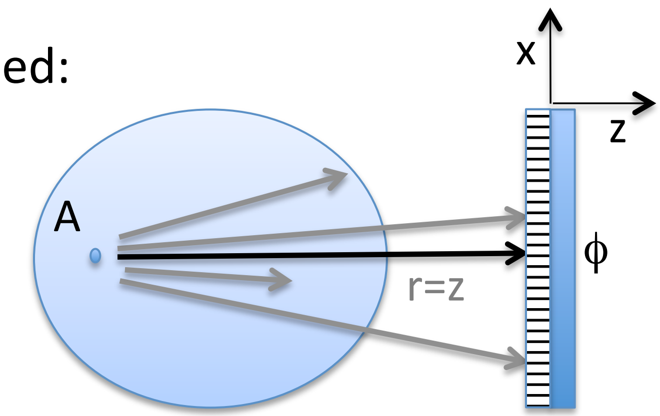
Different from X-ray imaging where r is fixed = source position



Add parallel collimator = consider only photons parallel to collimator holes

$$\phi_d = \frac{A}{4\pi z^2} e^{-\int_z^0 \mu(x,y,z';E)\,dz'} dz$$

The fluence on the detector at point (x,y) from a point like source at distance r=z from the detector

Imaging Equation

Photon fluence at (x,y) on the detector from a distributed source along z:

$$\phi(x,y) = \int_{-\infty}^{0} \frac{A(x,y,z)}{4\pi z^2}\, e^{-\int_{z}^{0} \mu(x,y,z';E)\,dz'}\, dz$$

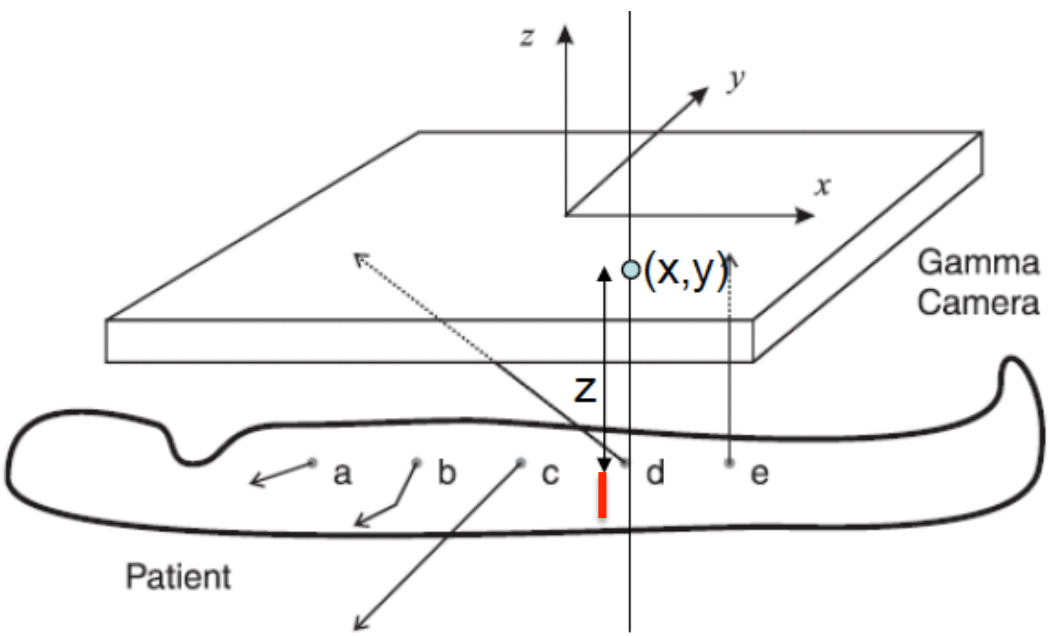
Simpler than radiography (or CT) since there is no integral over the energy spectrum (mono-energetic gamma emission)

But in scintigraphy (and SPECT) there are two sources of depth-dependent signal:

- Inverse square law
- Object-dependent attenuation

➔ consequence:

- Near activity is brighter
- Difference front/back



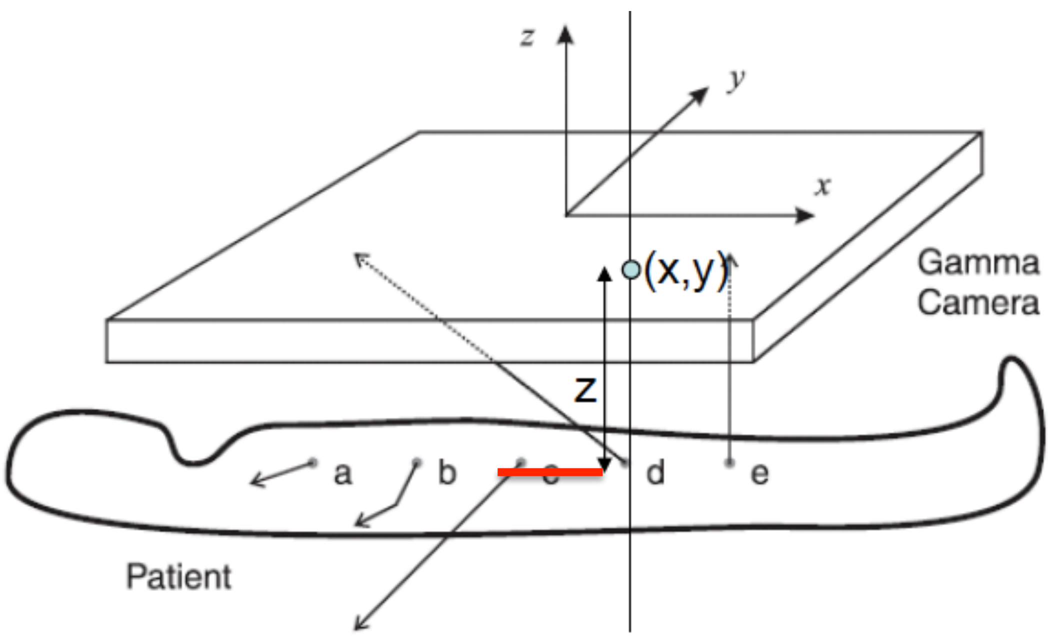
Simplified example: Planar source

Planar source in (x,y) at a fix z from the detector, $A(x,y,z)=A_{z0}(x,y)\delta(z-z_0)$:

$$\phi(x,y) = \boxed{\frac{A_{z_0}(x,y)}{4\pi z_0^2}}\, \boxed{e^{-\int_{z_0}^{0} \mu(x,y,z';E)\,dz'}}$$

Two factors reduce the photon fluence rate from the body:

- The inverse square law (independent on x,y)
- The integrate attenuation of photons (typically dependent on x,y)



Note: in the X-ray imaging case the source is fixed at $z_0 > z_{body}$ and the activity is replaced by the photon flux from the X-ray source.

Linear Attenuation Coefficients of Biological Tissues

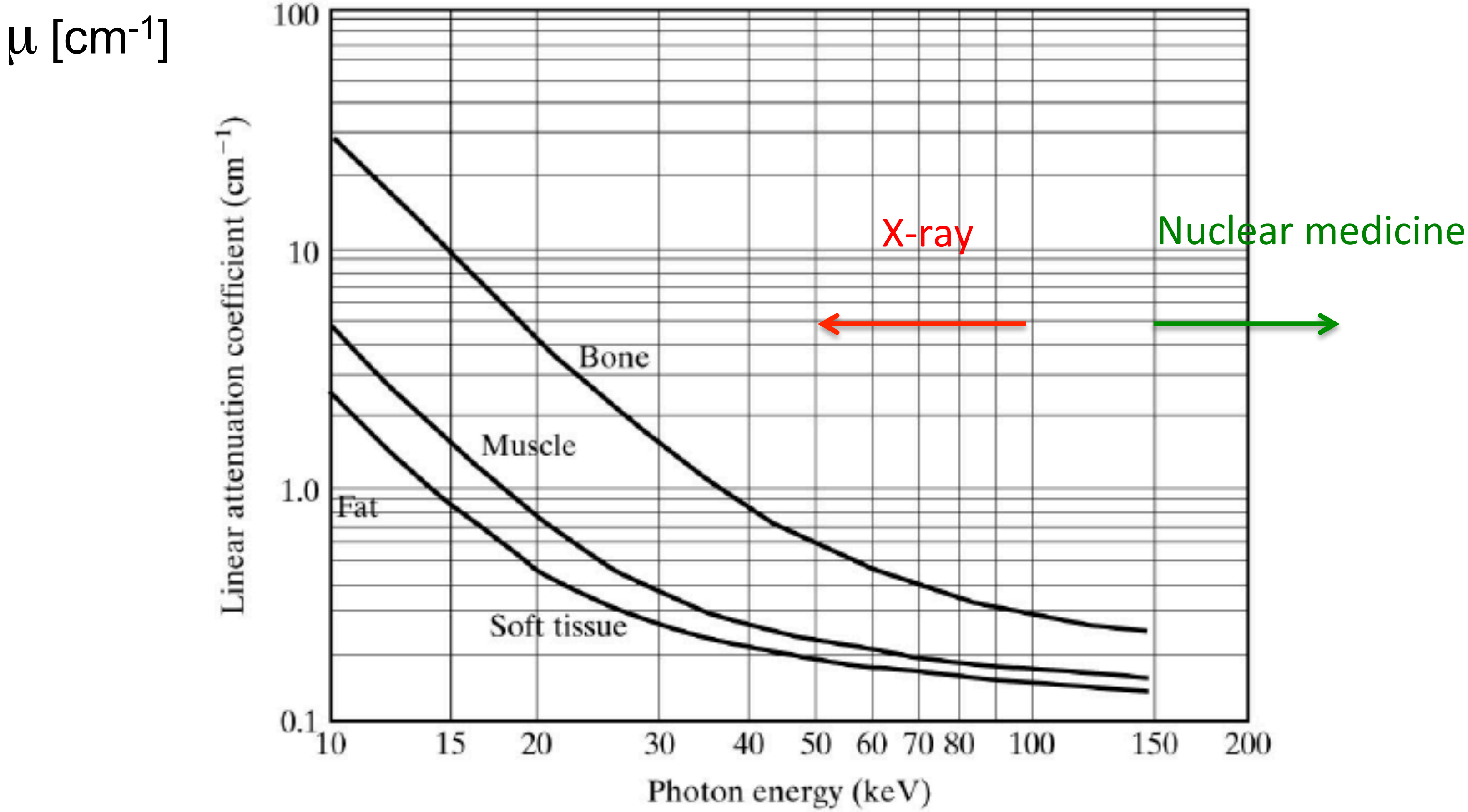
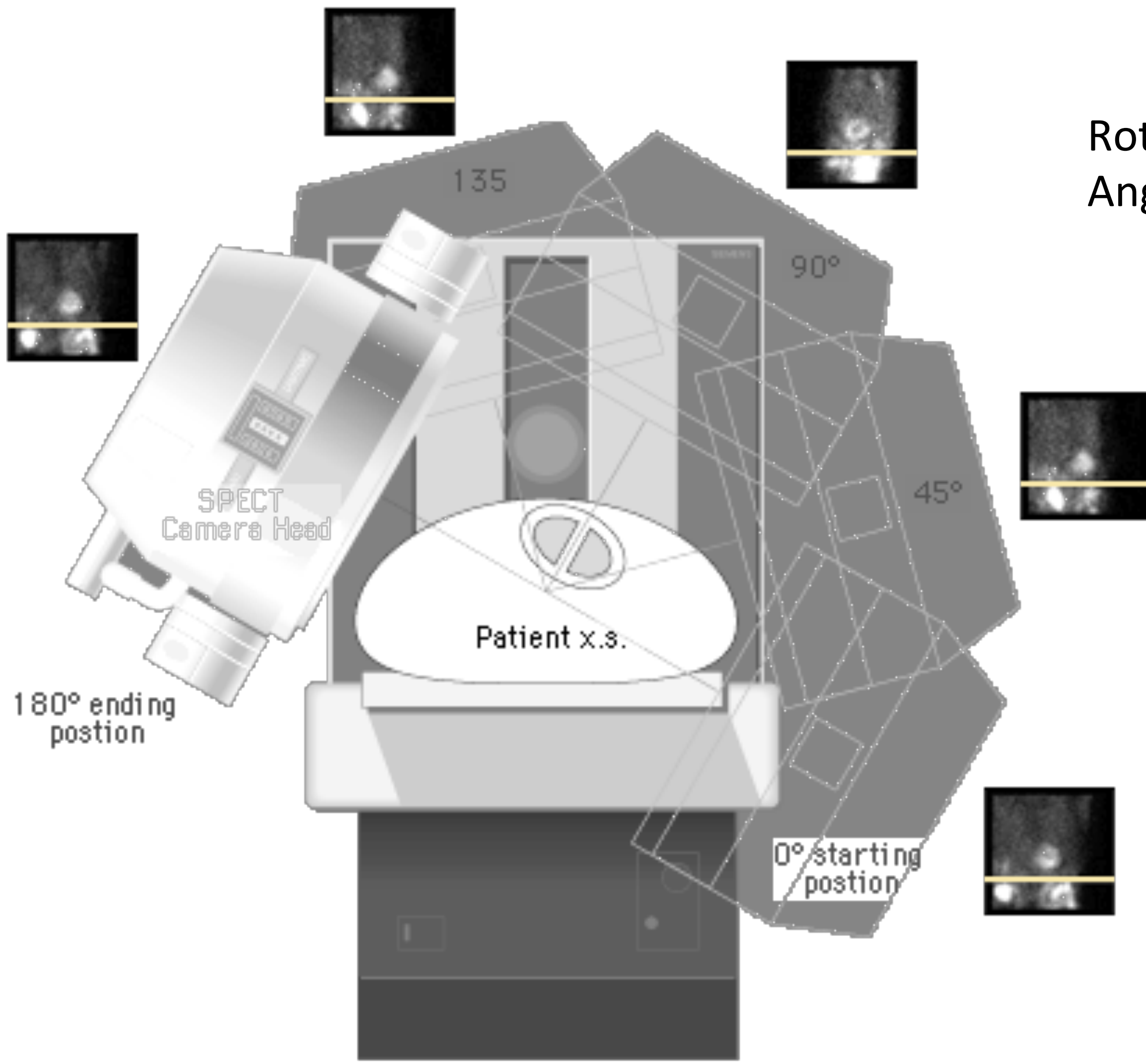


Figure 4.8

Medical Imaging Signals and Systems, by Jerry L. Prince and Jonathan Links.

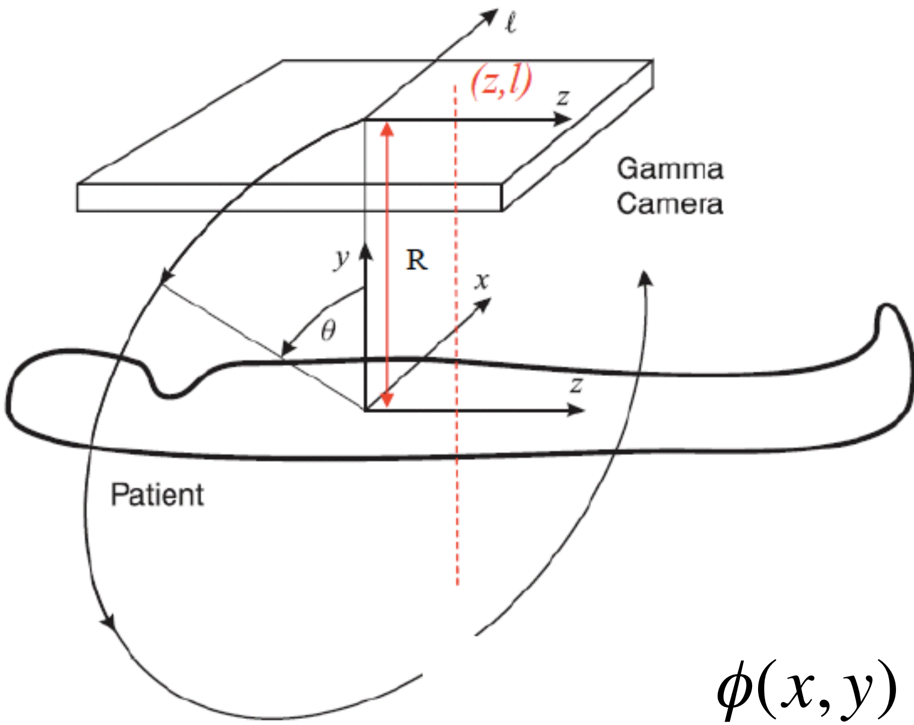
From 2D imaging to 3D tomography

Single-photon emission computed tomography (SPECT)



Rotating single (or multi)
Anger camera

Imaging Geometry and Assumption



Need two reference systems:
One fixed to the body, the second rotating with the detector

Photon fluence on the detector is the same as for scintigraphy but need to rename coordinates

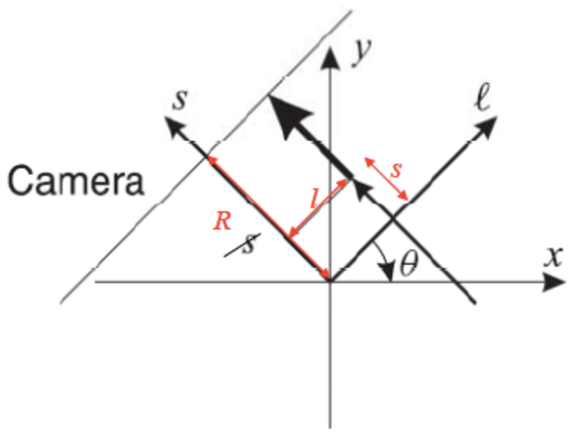
$$\phi(x,y) = \int_{-\infty}^{0} \frac{A(x,y,z)}{\cancel{4\pi z^2}} \cancel{e^{-\int_z^0 \mu(x,y,z',E)\,dz'}} \, dz$$

Practical approximation:
- Ignore the inverse square law
- Ignore the attenuation term

Works reasonably well in practice

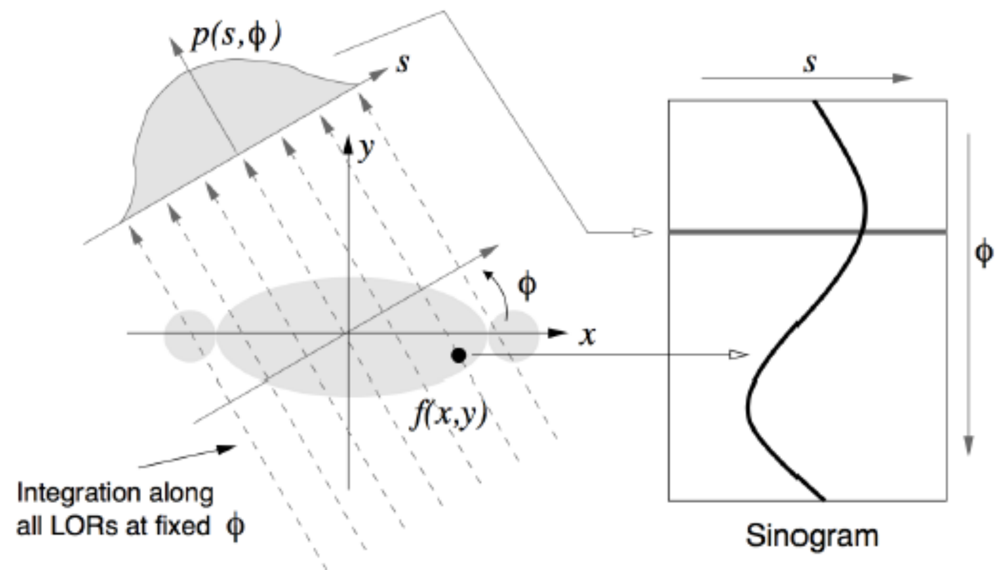
$$\phi(l,\theta) = \int_{-\infty}^{0} A(x(s),y(s)) \, ds$$

Image reconstruction



$$
\begin{aligned}
x(s) &= \ell\cos\theta - s\sin\theta \\
y(s) &= \ell\sin\theta + s\cos\theta
\end{aligned}
$$

Definition of sinogram
see lecture 7 where
f(x,y) = A(x,y) and
p(s,ϕ)=Φ(l,θ)



$$\phi(l,\theta) = \int_{-\infty}^{0} A(x(s), y(s))\ ds = \int_{-\infty}^{0} A(x,y)\delta(x\cos\theta + y\sin\theta - l)\ dxdy$$

Radon transform

Example: Gated myocardial perfusion SPECT

A ECG-gated tomographic acquisition represents a variation of a conventional perfusion SPECT

8 frames per cardiac cycle is considered a good compromise between temporal resolution, count density and acquisition time

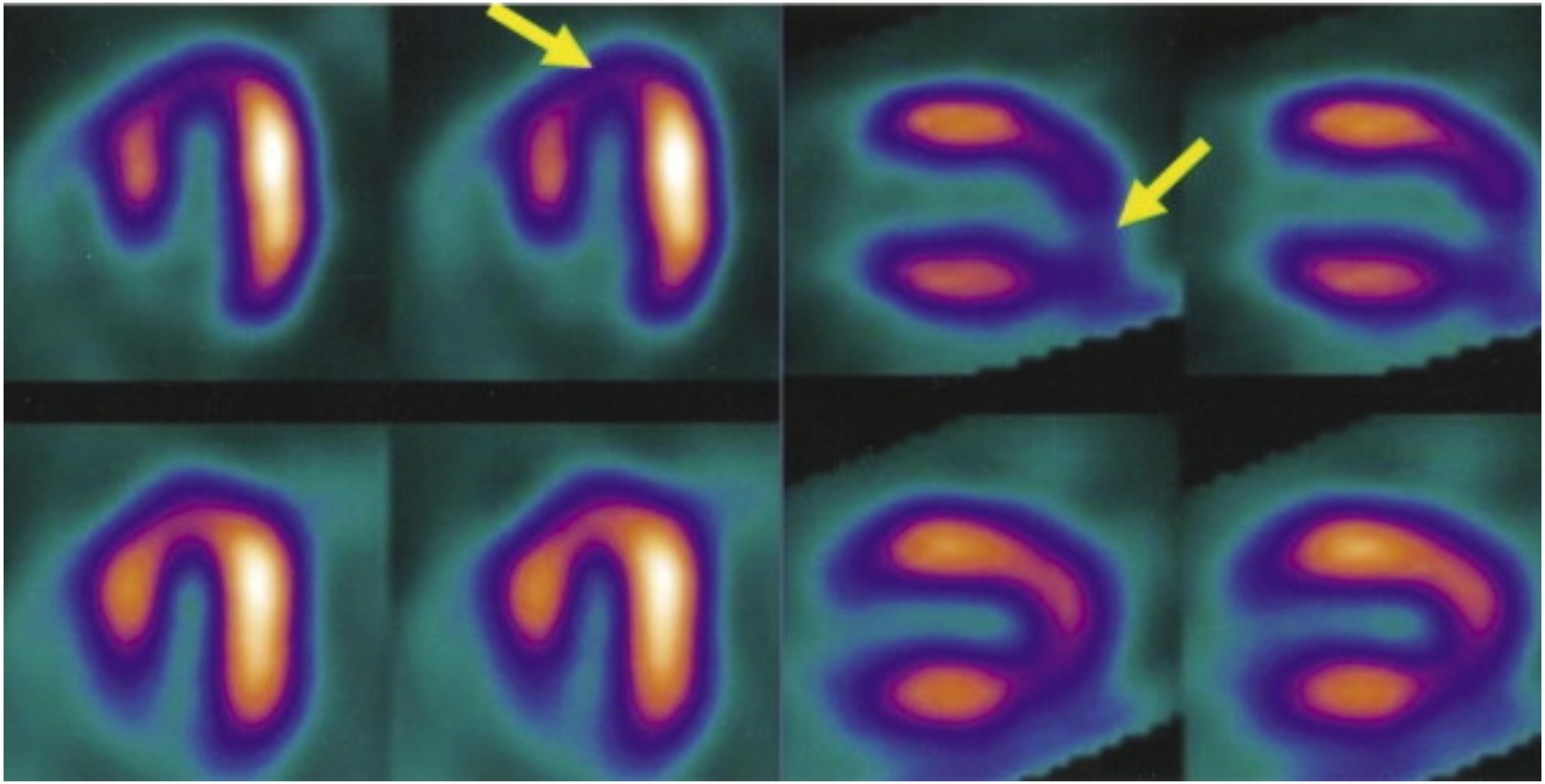
SPECT study consists of 32 angular steps, then the final file will contain 32 x 8 = 256 frames

Table 1.- Imaging guidelines of the American Society of Nuclear Cardiology.
TWO-DAYS PROTOCOL WITH ^{99m}Tc-MIBI

	Stress
Dose	20 - 30 mCi
Position	Supine
Waiting time	
Injection ===> Imaging	15 min - 1 h
Acquisition	
Energy window	20% symmetric
Collimator	LEHR
Orbit	180º (RAO-LPO)
Type of orbit	Circular
	Elliptic
Pixel size	6.4 ± 0.2 mm
Acquisition type	Continuous (non-gated)
	Step-and-shoot
Nº of projections	64
Matrix	64 x 64
Zoom factor*	1x - 2x
Time / projection	20 seg
Total time	25 min
ECG gating**	Yes
Frames / cycle	8

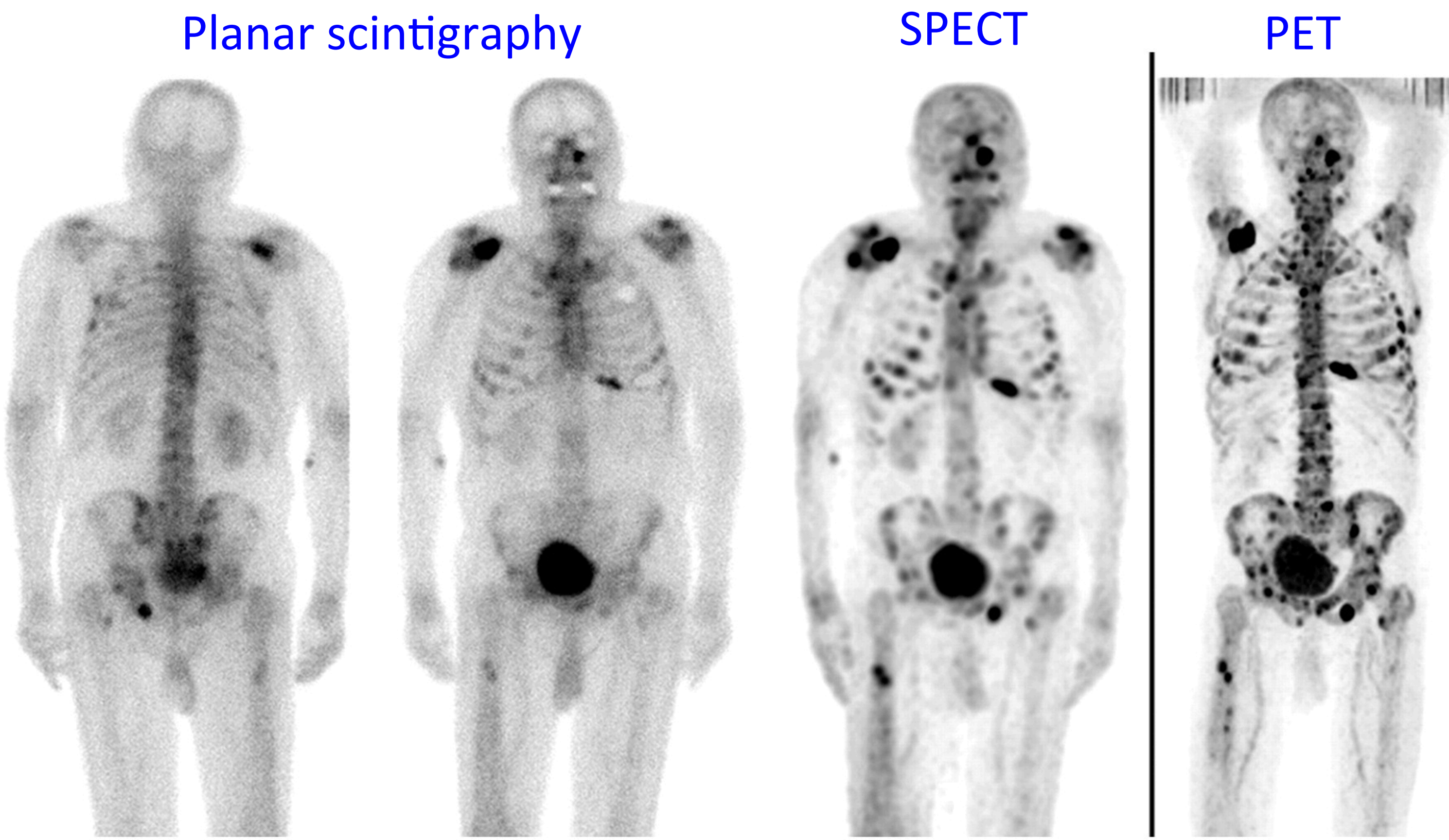
Note: a patient with regular beats occurring at least 80% of the time is considered adequate for gated acquisition.

myocardial perfusion SPECT



stress (top row) and rest (bottom row) SPECT mycrocardial perfusion test shows a reversible perfusion defect in the apex and the antero-apical region (arrows) consistent with myocardial ischemia. The color convention used shows normal perfusion by brighter colors and decreased perfusion by darker colors

Planar scintigraphy vs SPECT



From left to right: posterior and anterior ^{99m}Tc-MDP planar scintigraphy, ^{99m}Tc-MDP multiple-field-of-view SPECT, and ^{18}F-fluoride PET of 82-y-old patient with numerous bone metastases. In this patient, more lesions are typically detected by SPECT than by planar imaging, and ^{18}F-fluoride PET detects more lesions than does SPECT.

Summary

- Anger camera is the basic detector component for planar scintigraphy and SPECT
- Collimator and energy rejection to select proper line of response
- To obtain a tomographic reconstruction (SPECT) the anger camera is rotated around the patient and the projections at each angle are reconstructed with image reconstruction techniques (FBP)
- To solve analytically the FBP equation requires two important approximations in the definition of photon flux (…, …)
- SPECT delivers in most of the cases a more precise image than planar scintigraphy (but … longer acquisition times, higher cost)