Nuclear Medicine Physics

Lecture 8
Positron Emission Tomography

Mi-Ae Park
(miaepark@bwh.harvard.edu)

References :
- Physics in Nuclear medicine, 4th ed. by Cherry et al.
- Essentials of Nuclear Medicine Physics and Instrumentation, 3rd ed. by Powsner

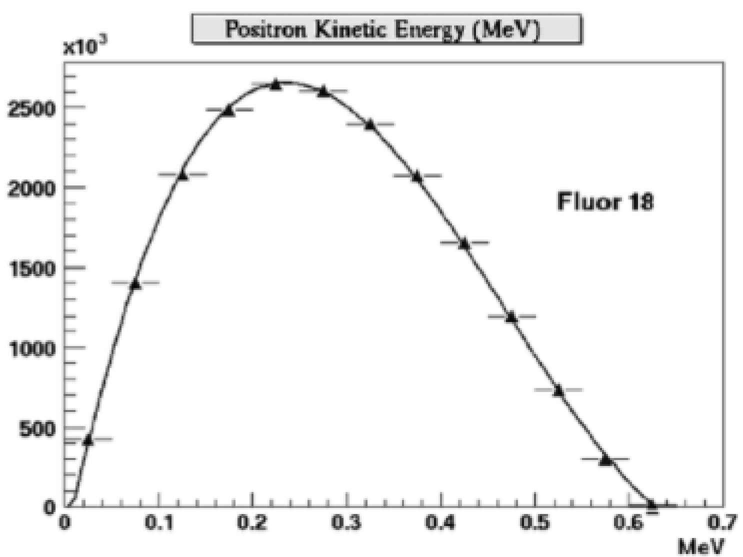
Positron Emission

- Too many protons
  - ✓ proton decays by positron (β^+) emission

$$p \rightarrow n + \beta^+ + \nu \qquad {}^{18}_{9}F_9 \rightarrow {}^{18}_{8}O_{10} + \beta^+ + \nu \qquad 97\%$$

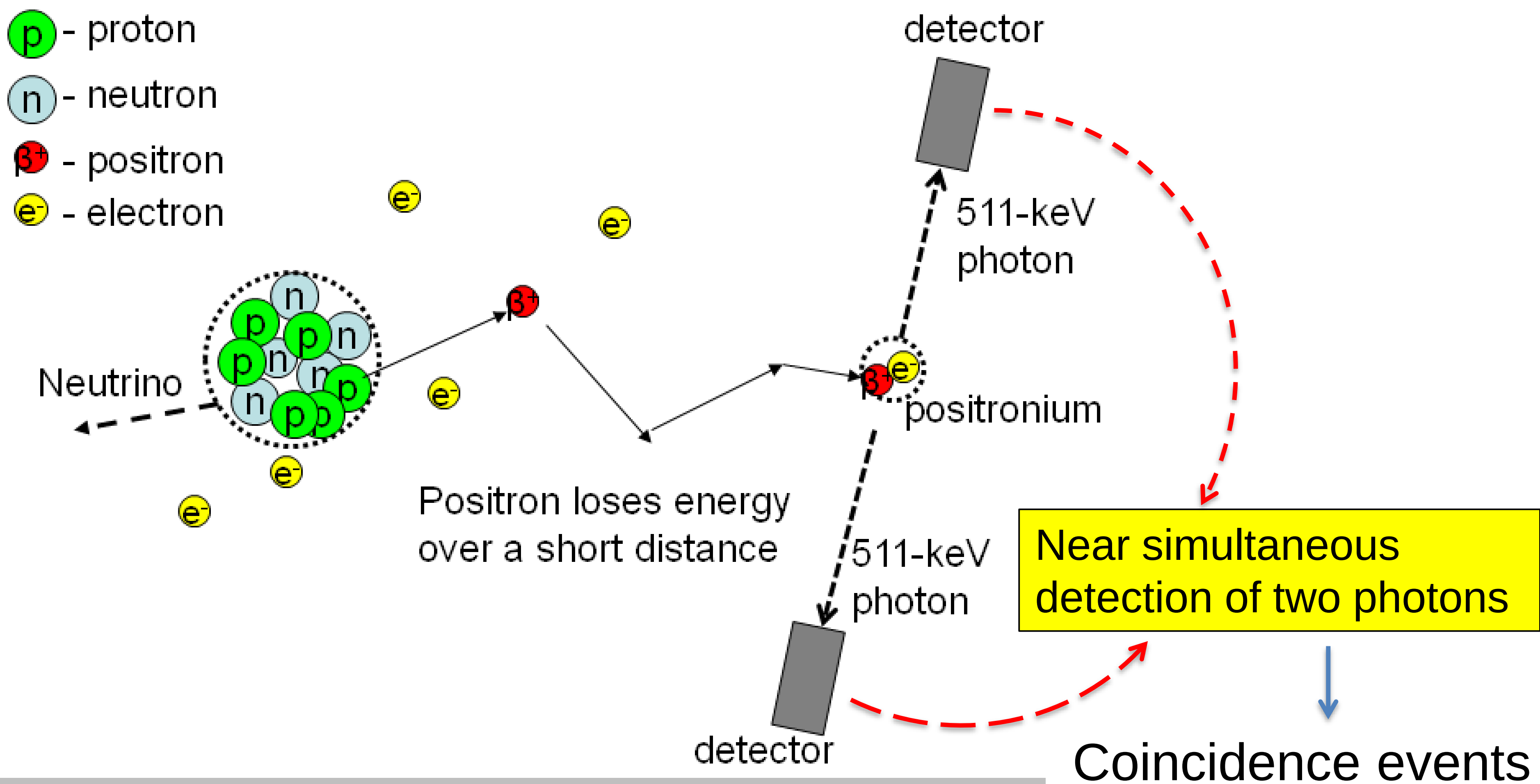
  - ✓ Electron Capture (EC)

$$p + e^- \rightarrow n + n \qquad {}^{18}_{9}F + {}^{0}_{-1}e \rightarrow {}^{18}_{8}O + n \qquad 3\%$$



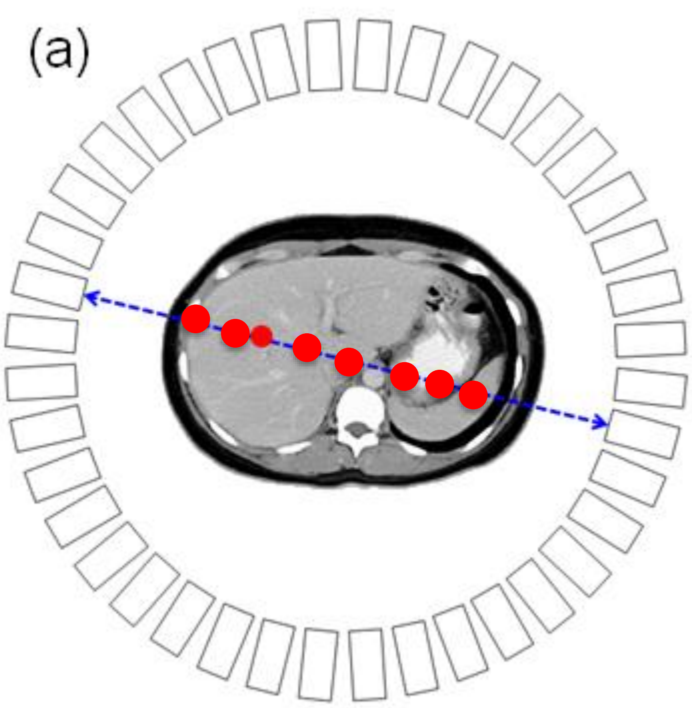
radionuclide	half-life	E_{max}(MeV)
Nitrogen-13 (${}^{13}_{7}N$)	9.97 m	1.2
Fluorine-18 (${}^{18}_{9}F$)	109.7 m	0.64
Rubidium-82 (${}^{82}_{37}Rb$)	1.27 m	3.15
Carbon-11 (${}^{11}_{6}C$)	20 m	0.97

Positron decay and annihilation

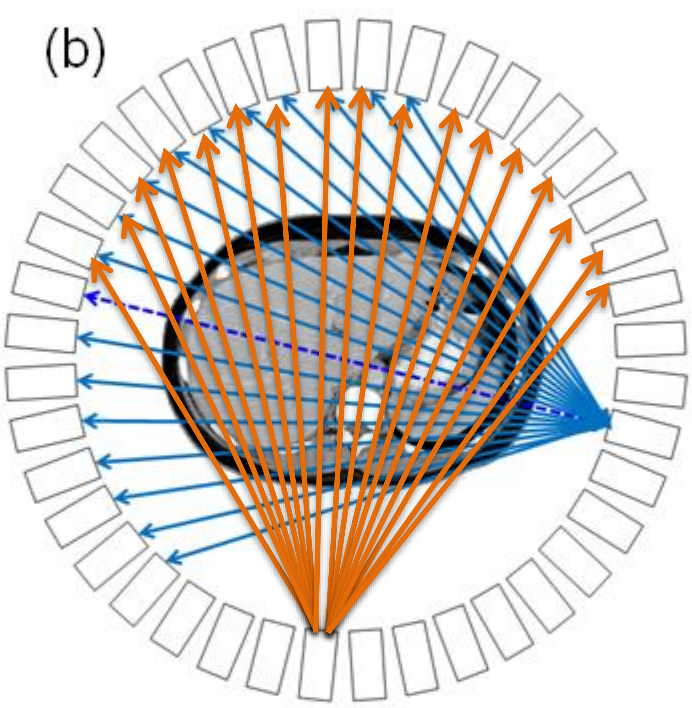


- ✓ Conversion of rest mass (β^+ & e^-) into radiation (511keV), **$E = mc^2$**
- ✓ The positrons themselves are not directly detected to make images

Detector Geometry for Clinical PET system



line of response (LOR)



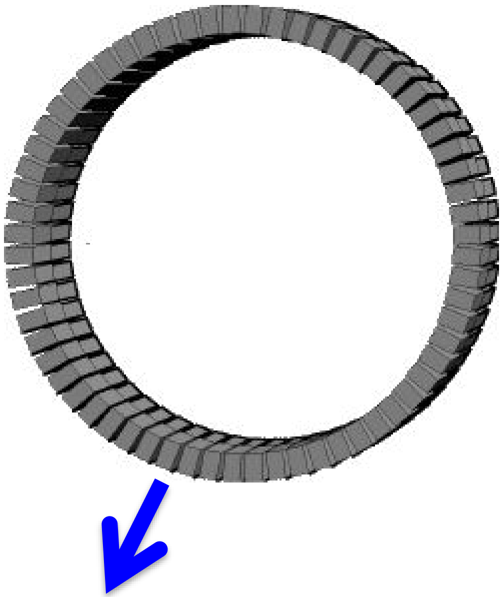
LORs which connected a single detector to all the detectors on the other side of the patient

Each LOR will have recorded a total number of counts proportional to the integral of the radioactivity concentration along that LOR.

Detector Geometry for Clinical PET system

Modern PET scanners :

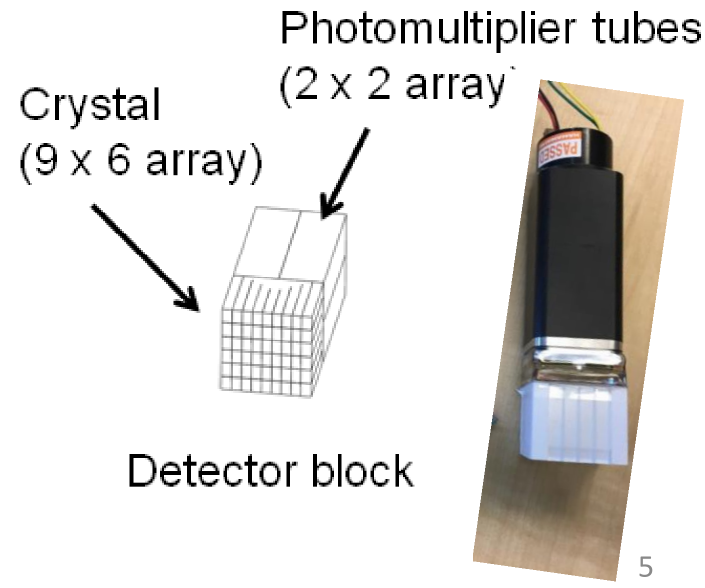
- many more rings of detectors surrounding the patient
- permit more slices within a larger axial range (typically 15-20 cm) to be imaged simultaneously.



For example, GE DRX PET scanner

- 24 rings with 630 LYSO crystals per ring

  Total ~ 15,120 crystals
- Crystal dimensions : 4.2 x 6.3 x 30 mm^3
- Axial range : 15.2 cm



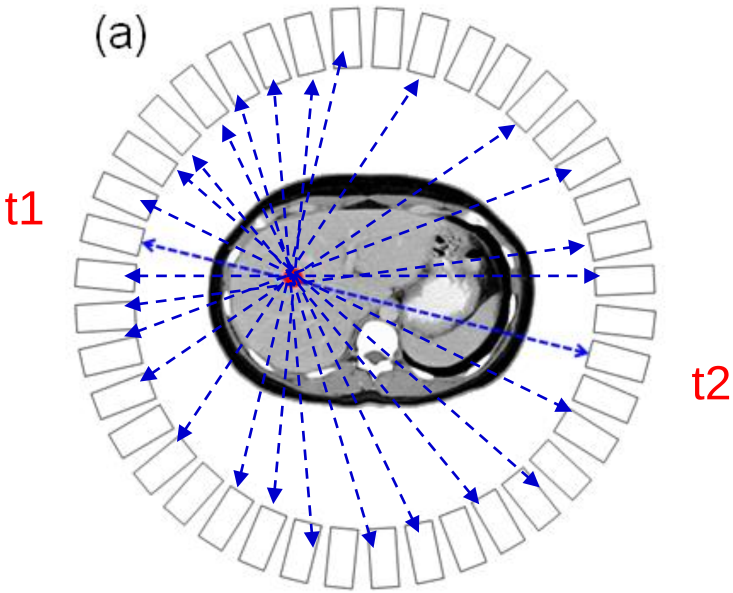
Some properties of scintillation crystals

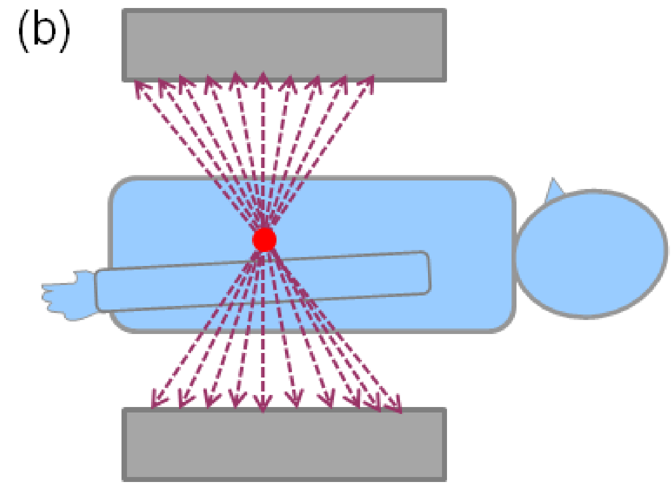
Property	NaI(Tl)	BGO	LSO/LYSO	$LaBr_3$
Density (g/mL) (for stopping)	3.67	7.1	7.4	5.3
Decay time (ns) (timing window)	230	300	40	35
Photo fraction (%)	100	25	45	<20

- Annihilation-photon interacts in the crystal
  → light flash take place almost immediately
  → Determine if a coincident photon was detected at almost the same time on both detectors

- too few light photons→ a large uncertainly in determining the interaction site → degrades spatial resolution (and energy resolution)

Coincidence events and line-of-response (LOR)





If (t1 - t2) < T, accept the coincidence event

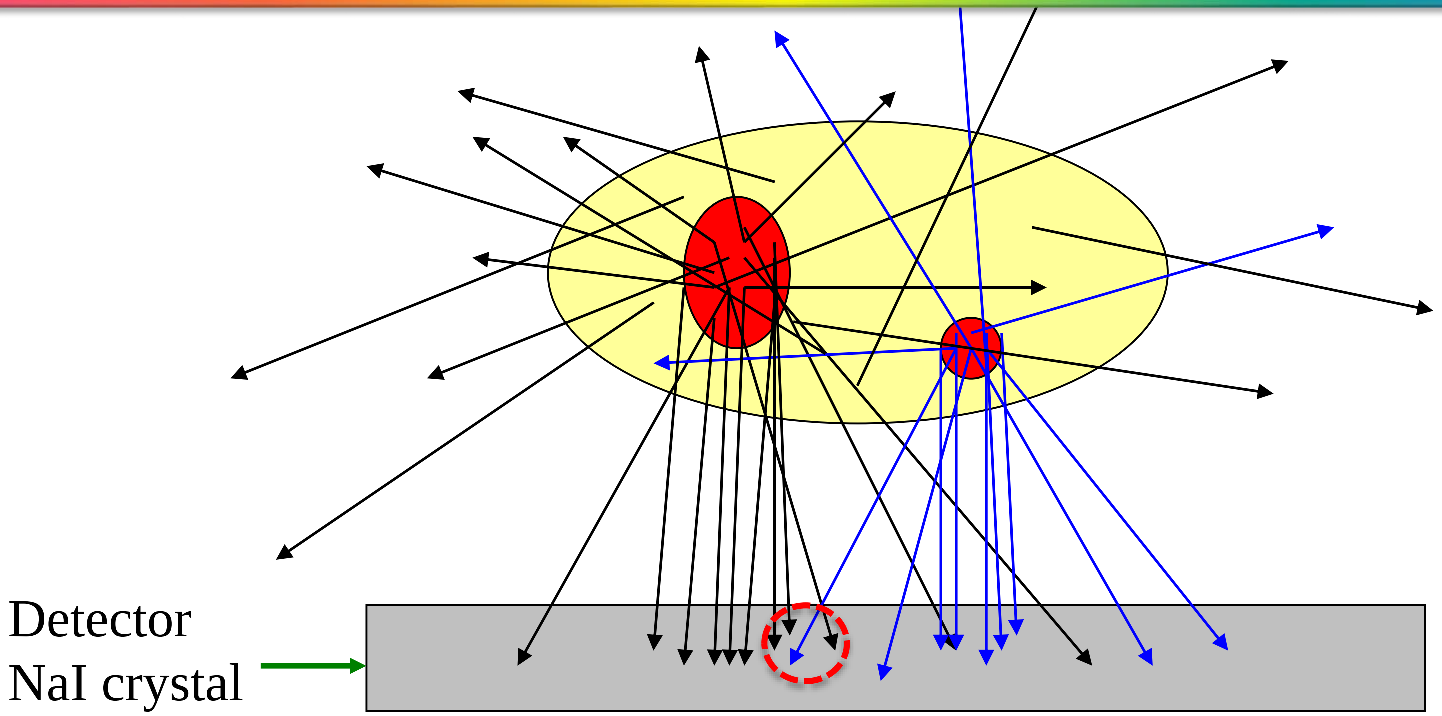
Many coincidence events are detected in 3D

T : Coincident timing window

Example
T = 9.5 nsec for BGO
T = 4.9 nsec for LYSO

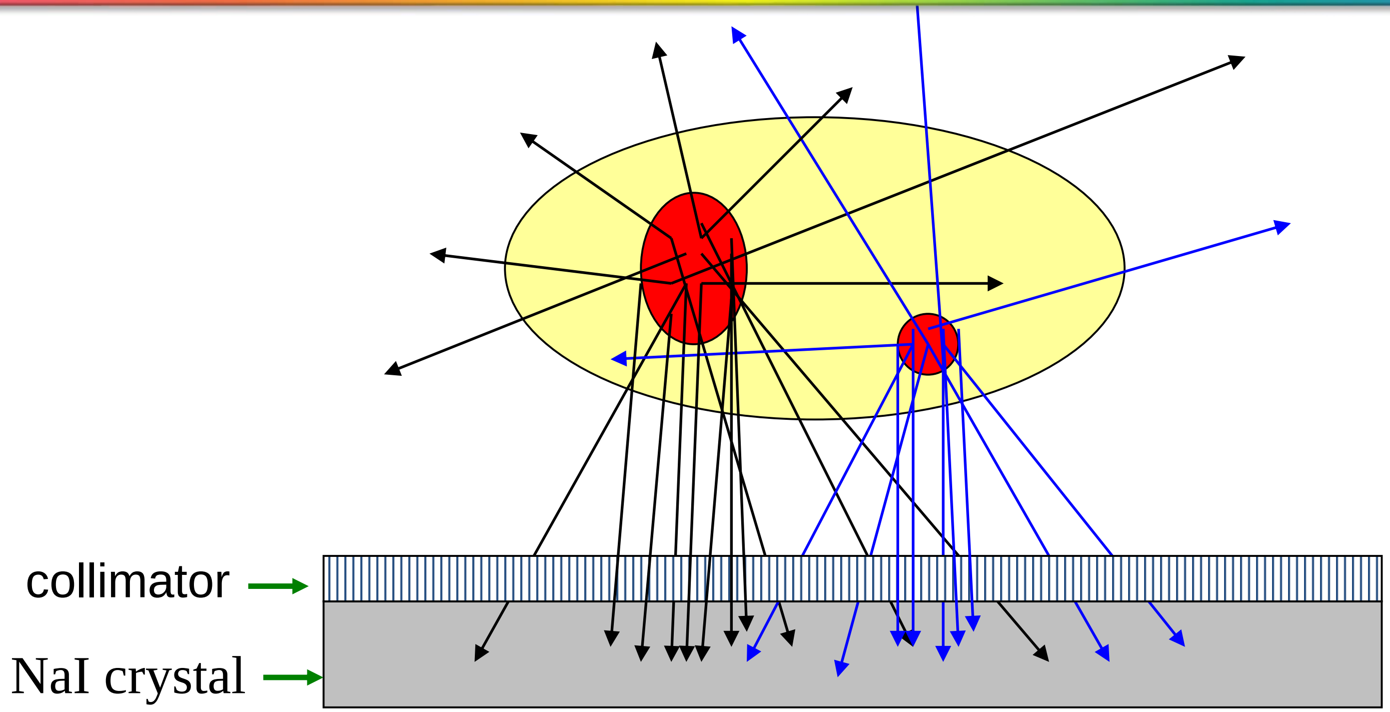
(electronic collimation)

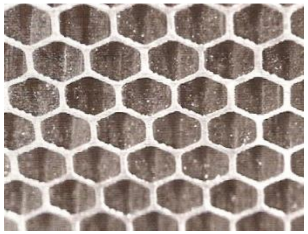
Collimator in SPECT imaging



- Gamma rays are emitted to any direction in space.
- The detected gamma rays can come from any location in the body

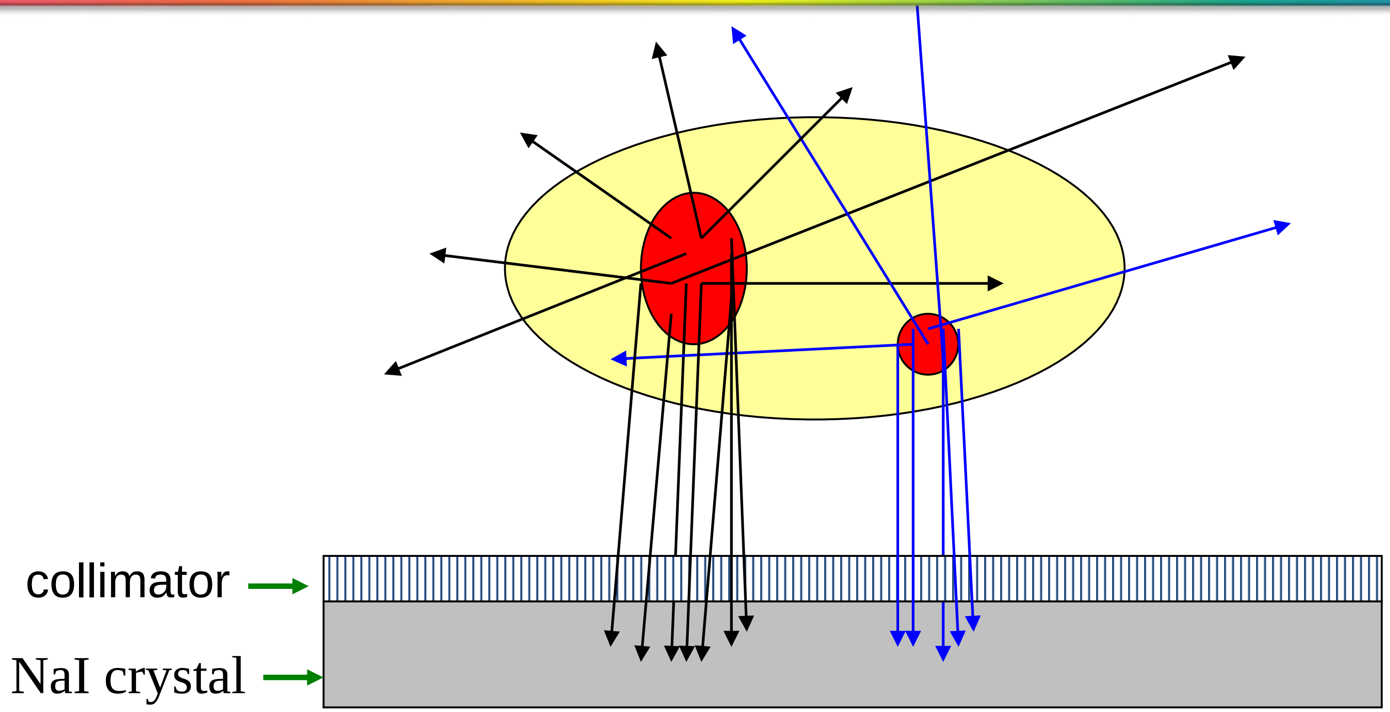
Collimator in SPECT imaging

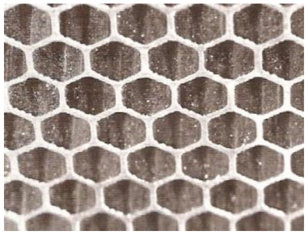




- The collimator establishes a one-to one correspondence between locations on the detector and those within the organ.

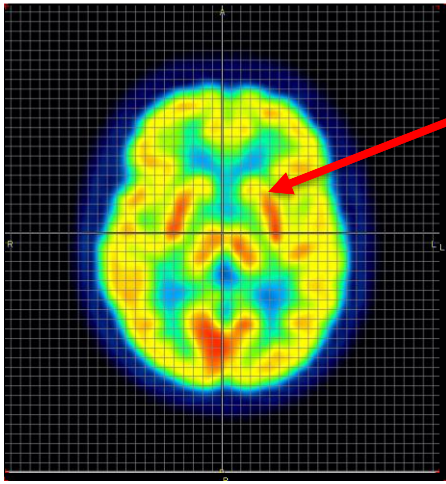
Collimator in SPECT imaging





- The collimator establishes a one-to one correspondence between locations on the detector and those within the organ.
- The collimator allows those gamma rays traveling along certain directions to reach the detector (only a few out of 10,000 for parallel collimation).

PET images



FDG brain PET

Voxel values in radioactivity concentration, Bq/ml or kBq/ml

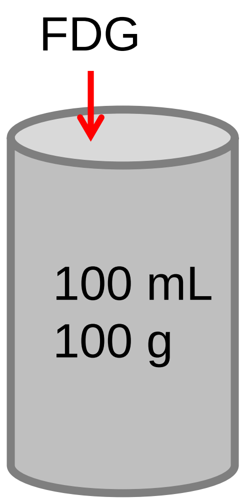
- administered dose
- patient (size, disease, etc)
- Properties of radiotracer
- Imaging time after injection
- scanner-specific calibration
- reconstruction

$$\mathrm{SUV\ (g/ml)} = \frac{\mathrm{Tissue\ \ activity\ concentration\ } (\frac{\mathrm{kBq}}{\mathrm{ml}})}{\mathrm{Injected\ dose\ (kBq)\ /Patient\ weight(g)}}$$

Standardized Uptake Values (SUV)

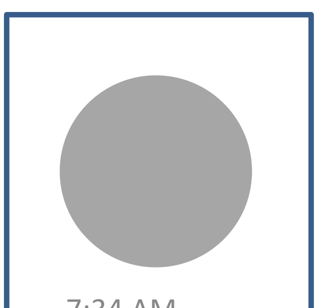
The simple property of a uniform distribution, no activity excreted, assuming a tissue density of 1 g/cm^3

Original PET images : radioactivity concentration in units Bq/mL



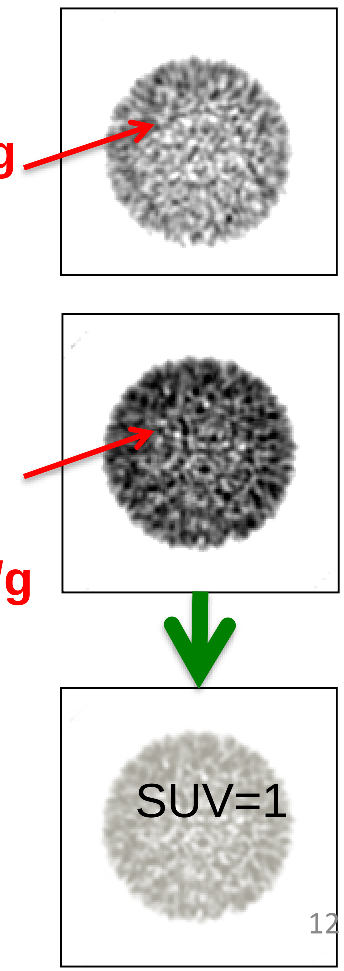
- Add 100 Bq FDG, Concentration = 1 Bq/mL
- Normalized to weight : 100Bq / 100g = **1Bq/g**
- SUV = (1Bq/mL) / (1Bq/g) = **1g/mL**

Transverse view

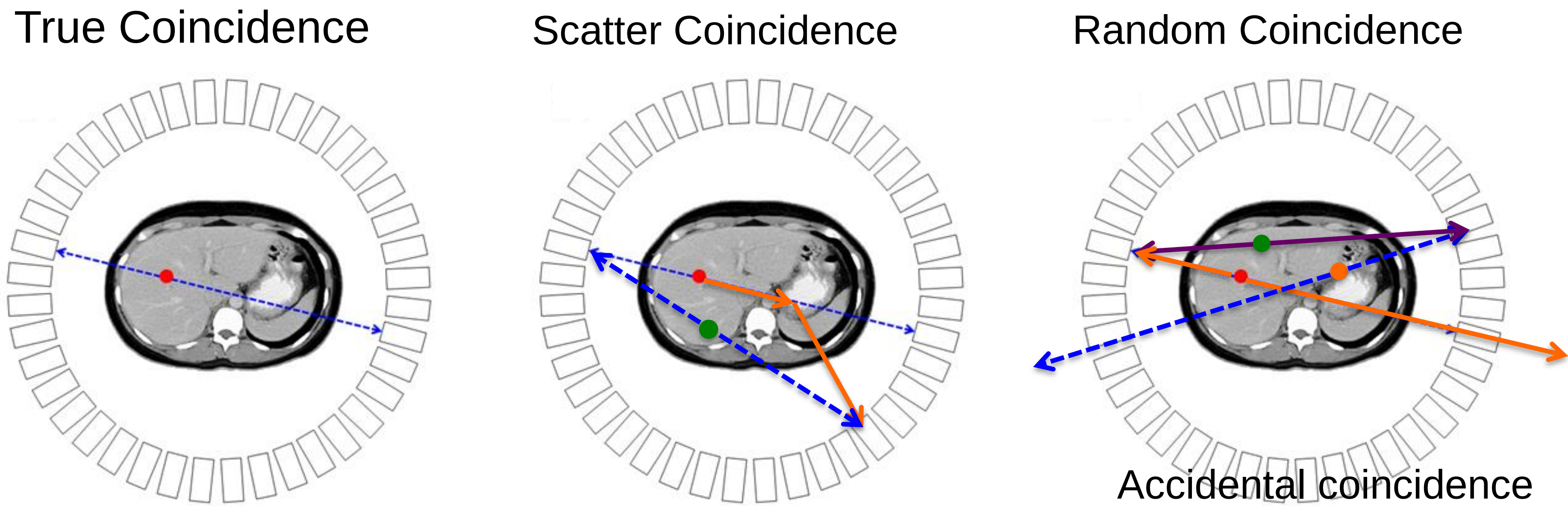


- Add 200 Bq FDG, Concentration = 2 Bq/mL
- Normalized to weight : 200Bq / 100g = **2Bq/g**
- SUV = (2Bq/mL) / (2Bq/g) = **1g/mL**

Transverse view



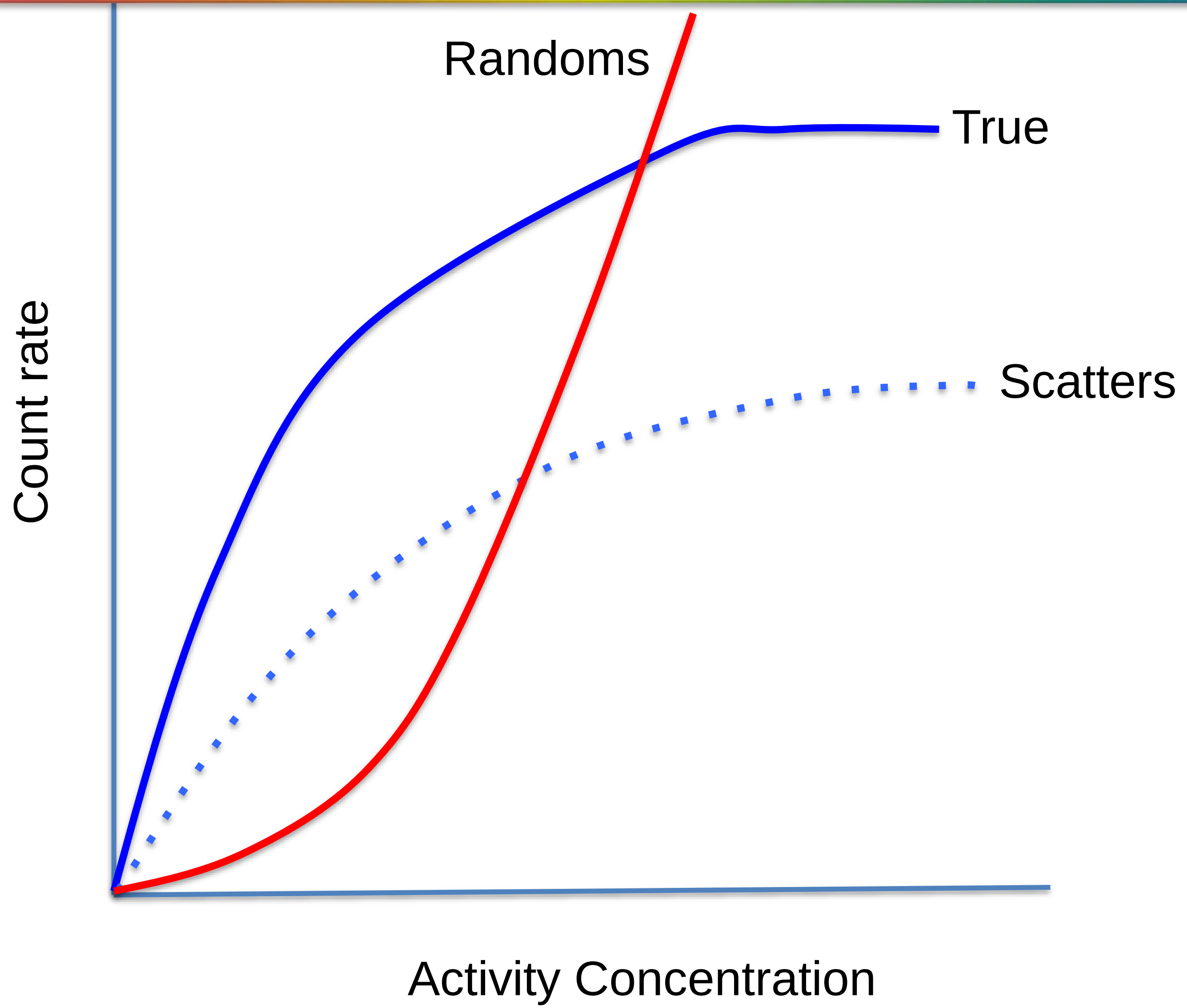
Coincidence Events



- T : one positron-electron annihilation are detected back-to-back along the correct LOR.
- S : one or both of the annihilation photons undergo Compton scattering in the patient before being detected
- R : two annihilation photons originating from two <u>different</u> nuclear decays to be detected by chance within the coincidence time window

In order to obtain quantitative imaging, corrections must be applied for randoms, attenuation, and scatter.

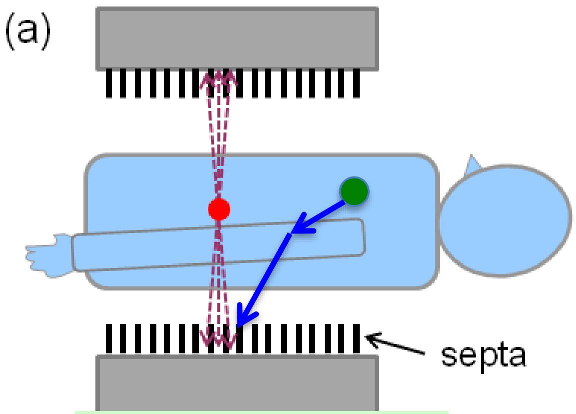
Coincidence Events



2D vs 3D for scatter correction

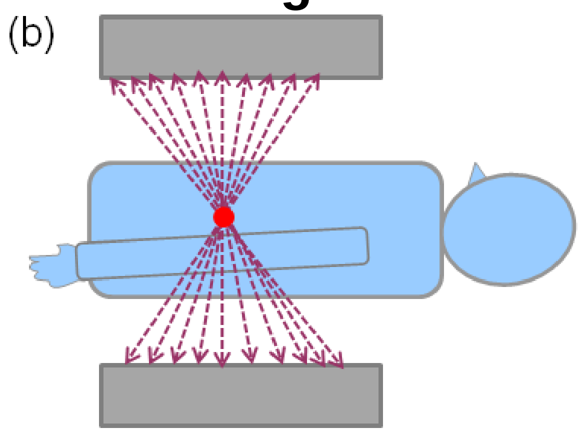
- Can be removed using anti-scatter grids, i.e., 2D acquisition
- Energy window (425 ~ 650 keV)

2D : with Grid

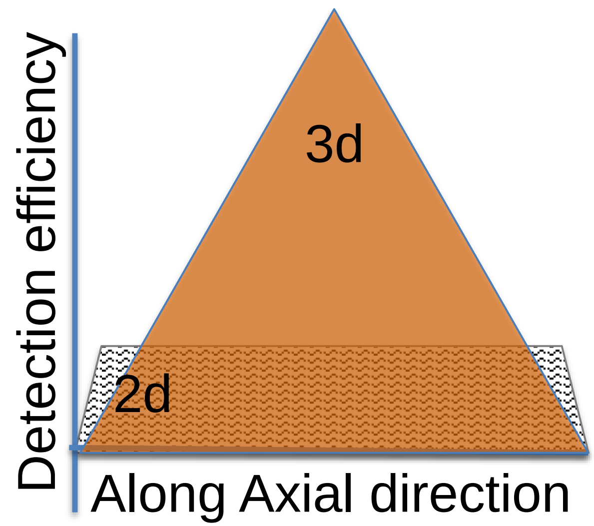


~ 15% scatters

3D : no grid

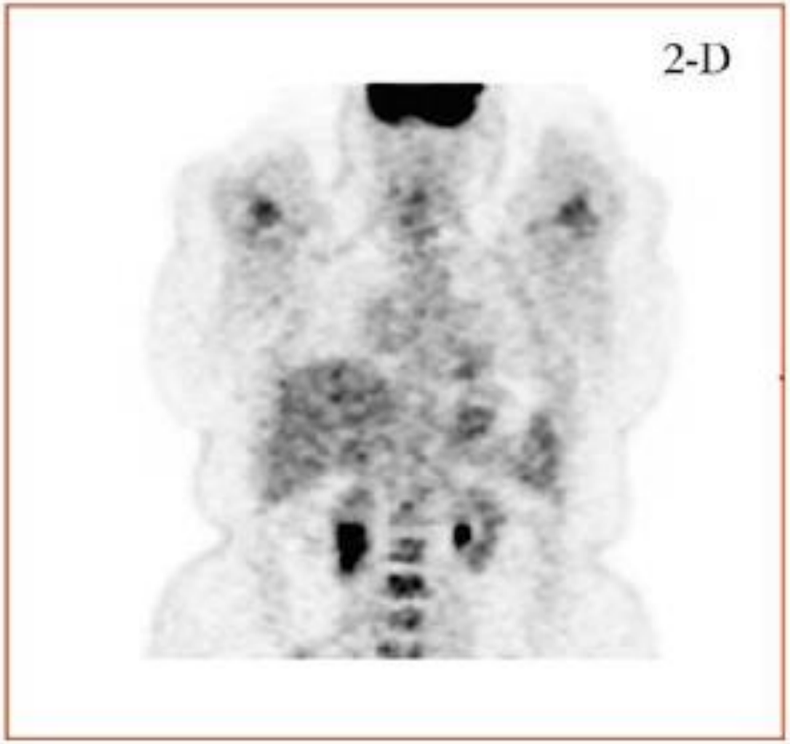


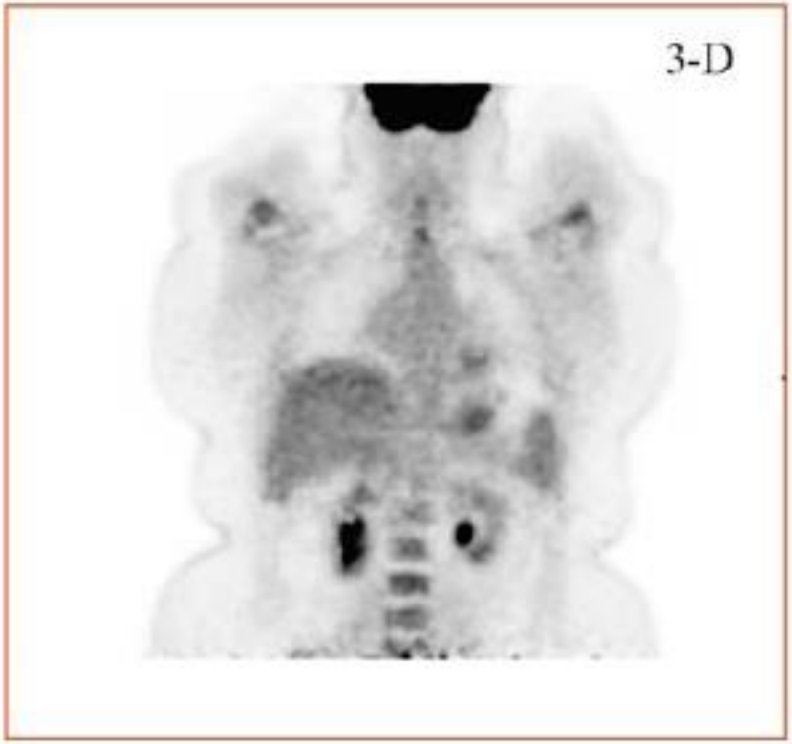
~ 40% scatters



- **<u>3D mode only for new scanners</u>**
  - ✓ fast and high capacity computers available
  - ✓ improved scatter & random corrections
  - ✓ fast electronic processing technique
  - ✓ high sensitivity (4-6 times greater),

Scan Type : 2D vs. 3D





For a large patient, PET images showed reduced noise level in the 3D image compared to the 2D image.

→ Less use of the anti-scatter grid

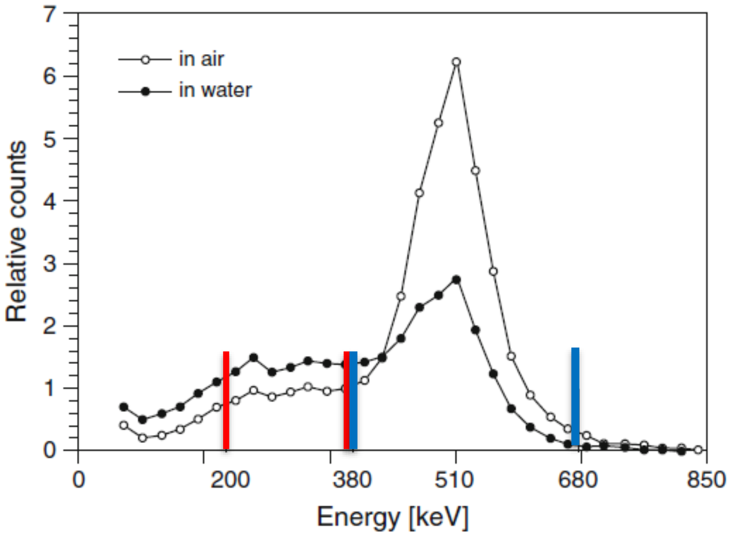
Scattered coincidence events

Scatter events arise from the same annihilation event and increase with

- ✓ energy window
- ✓ radioactivity in the patient
- ✓ size of the patient
- ✓ density of tissues and detector materials

→ increase the background counts
→ degrade the image contrast.

1. **Dual energy method**
   - photopeak window
   - low energy window



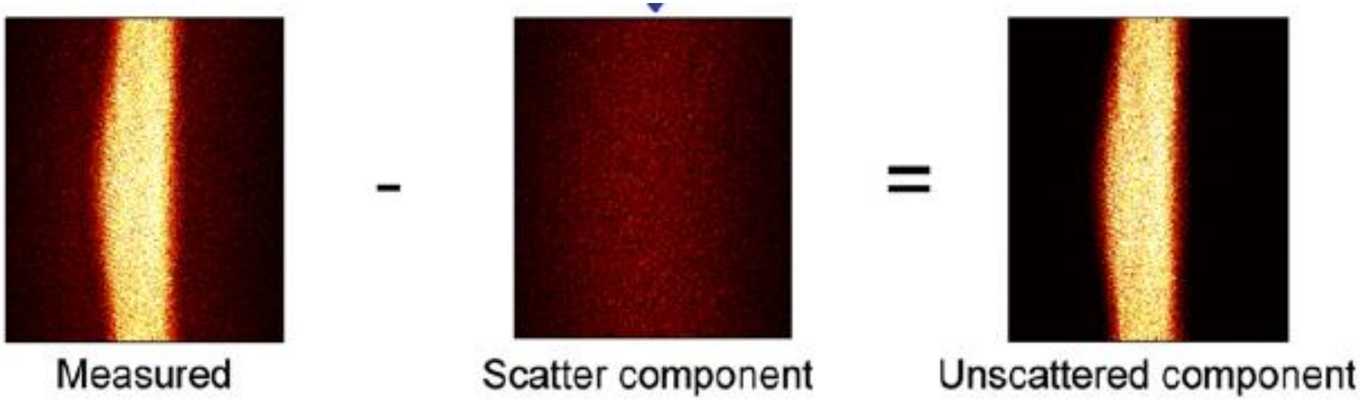
From Saha GB

Scatter correction

2. Model-based scatter-correction technique : most commonly used

- Utilize “preliminary PET” and transmission (CT) images
- Monte Carlo calculation of Compton scattering
- model of the scanner geometry and detector systems to calculate the percentage of photons falling on each detector, using the Klein-Nishina formula.
- Determine the magnitude and spatial distribution

MC simulated



H. Zaidi and M. Montandon, Scatter Compensation Techniques in PET

Random (accidental) coincidence events

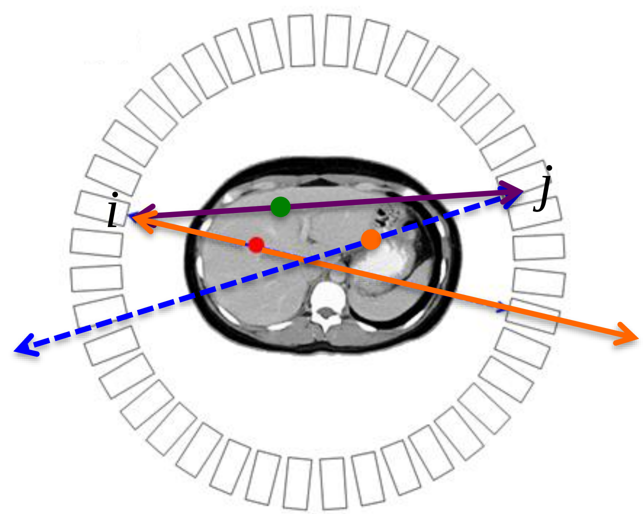
Randoms increase with energy window
coincidence timing window
radioactivity

- Randoms raise the background counts on the images
  → causing artifacts and loss of image contrast
- Random events vary with A^2
- the faster electronics and shorter time window minimize random events

1. Random correction from singles

- may be estimated from the count rate in individual detectors (<u>singles</u>)
- For crystal i and j, collect single events for timing window (T ~ 10ns)

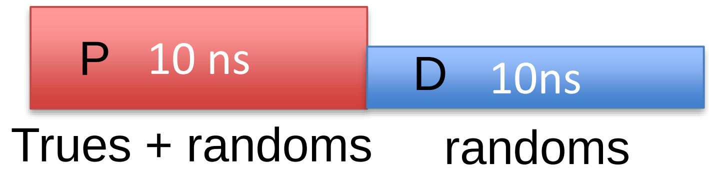
  random $R_{ij} = 2*T*S_i*S_j$



Random (accidental) coincidence events

2. delayed timing window.

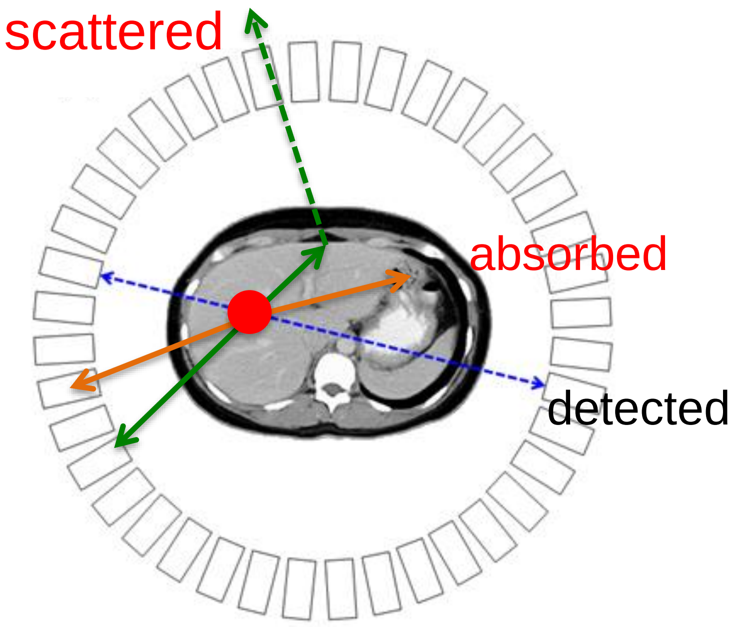
- After coincidence events (P) detection for 10ns, collect # of singles (D) for another 10ns.
- True events = P – D



- Shorter timing window will decrease the randoms
- BGO~10 ns, LYSO ~ 5ns → LYSO has less randoms
- LYSO based Time-of-Flight imaging will reduce the randoms.

Attenuation

- Photons are absorbed or scattered, and not being detected
- The number of detected photons << the number of emitted photons



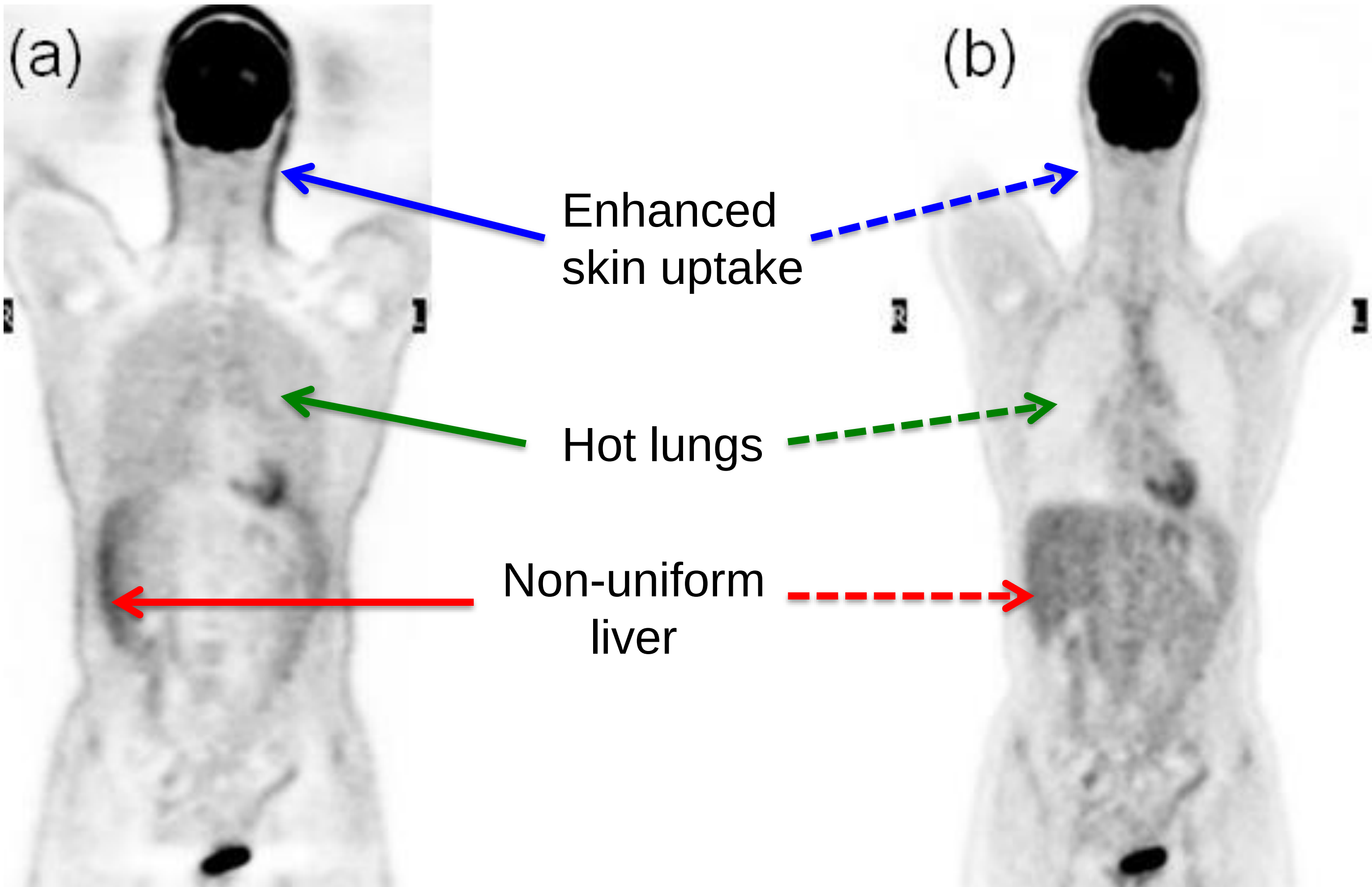
the lost photons need to be estimated to calculate the original number of emitted photons

→ attenuation correction

511keV photons,
HVL (tissue) ~ 75 mm (46 mm for Tc99m)
HVL (lead) ~ 4.2 mm (0.2 mm for Tc99m)

In order to obtain quantitative imaging, corrections must be applied for randoms, scatter, and attenuation

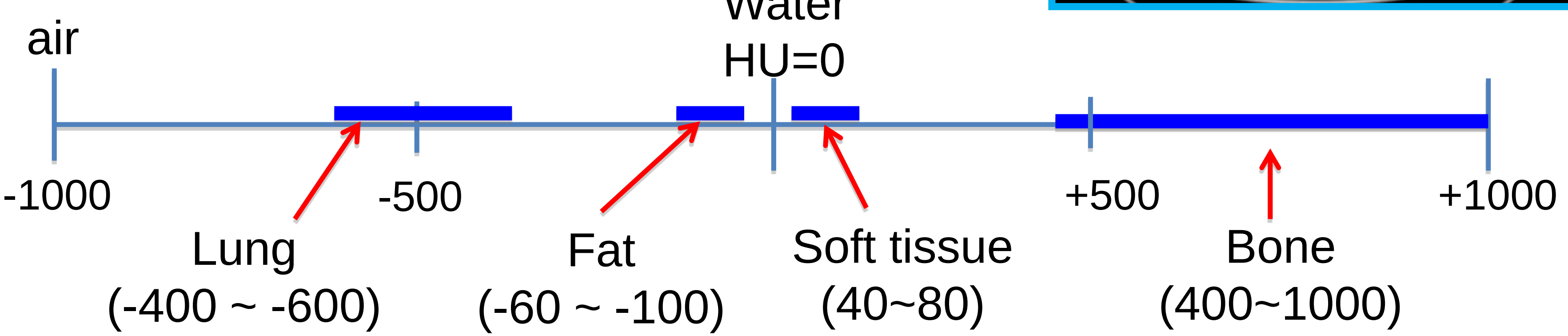
No attenuation correction | **With attenuation correction**



Normal FDG study

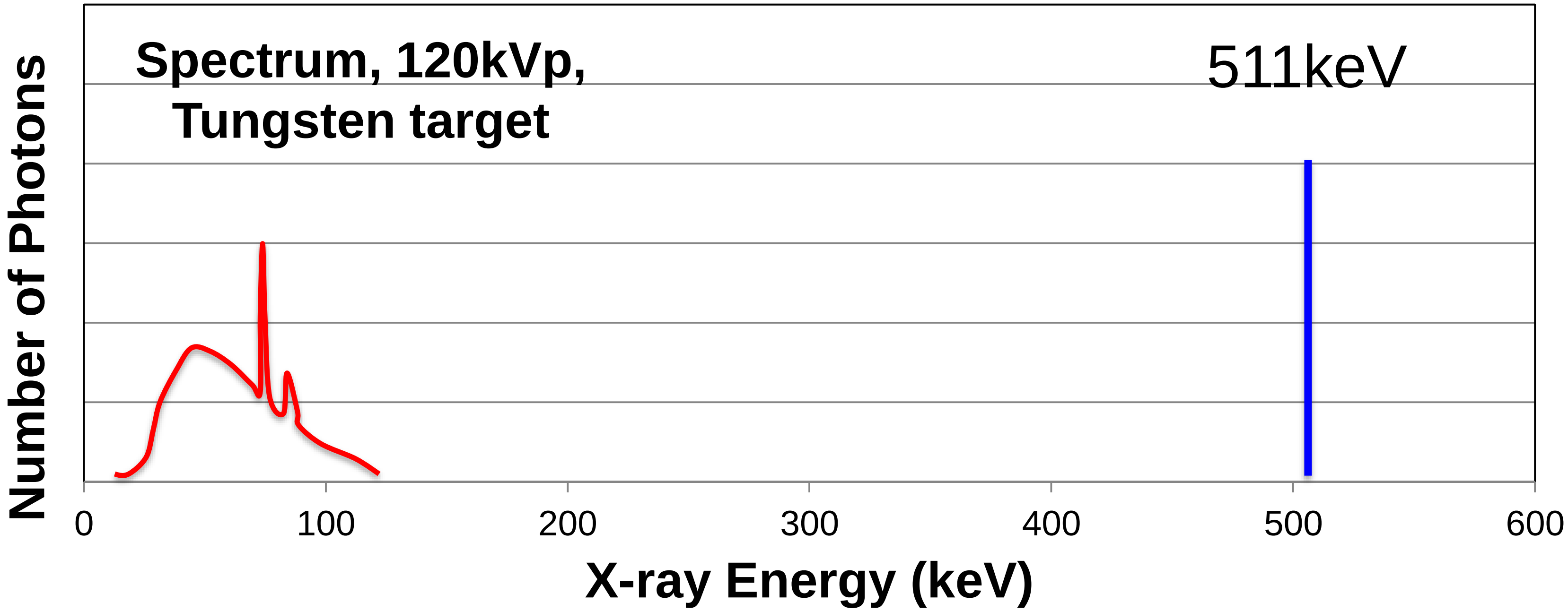
Replace the attenuation coefficient calculated for each voxel of the reconstruction matrix with an integer (CT number or HU)

$$\mathbf{HU} = \frac{\mu_{voxel} - \mu_{water}}{\mu_{water}} \times 1000$$



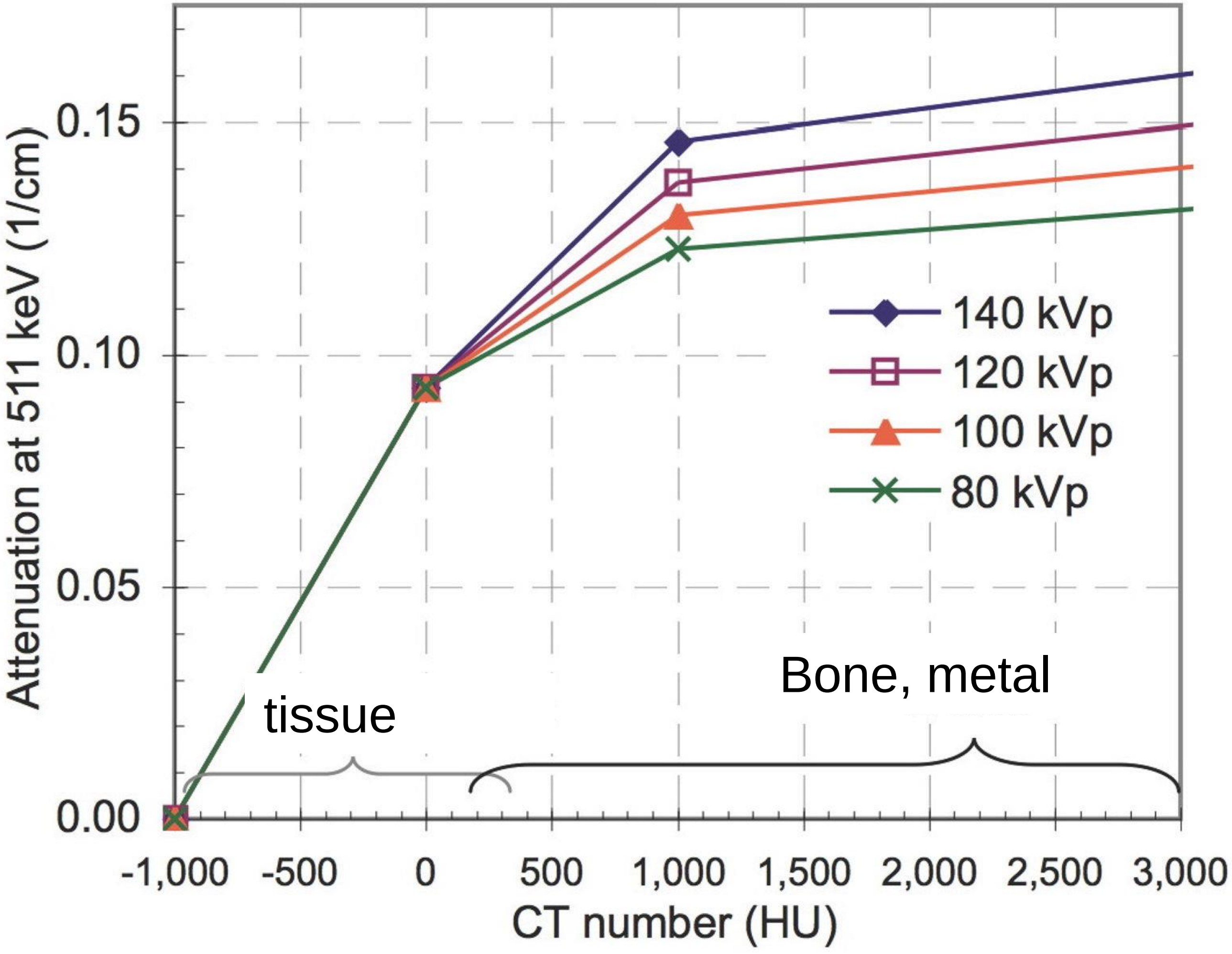
The attenuation coefficient, μ, is energy dependent, μ(E)

Bremsstrahlung spectrum (x-rays) vs. 511keV photons



Different attenuation coefficient for different energies

Conversion of HU to attenuation coefficient

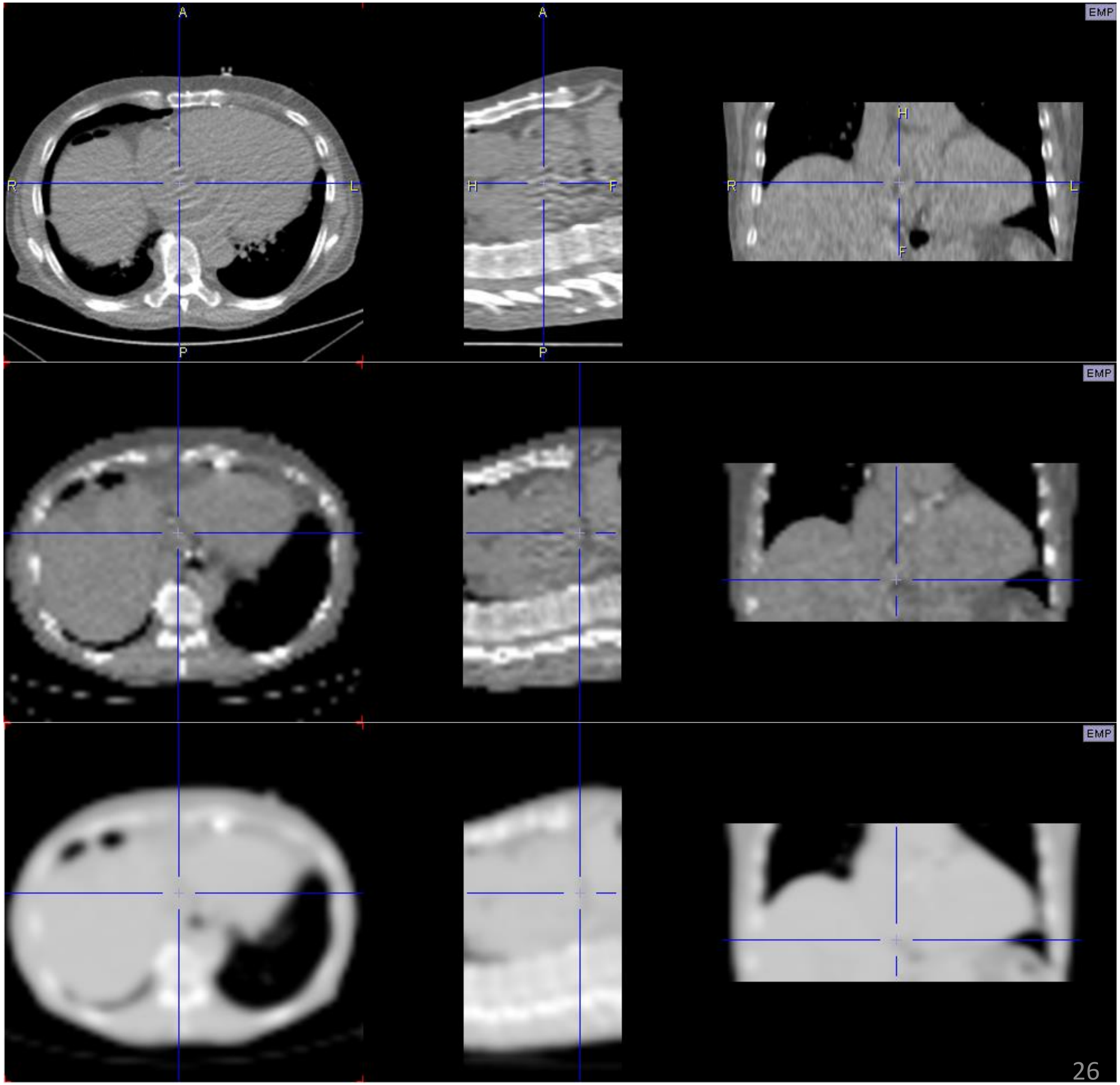


Conversion of CT images to attenuation map

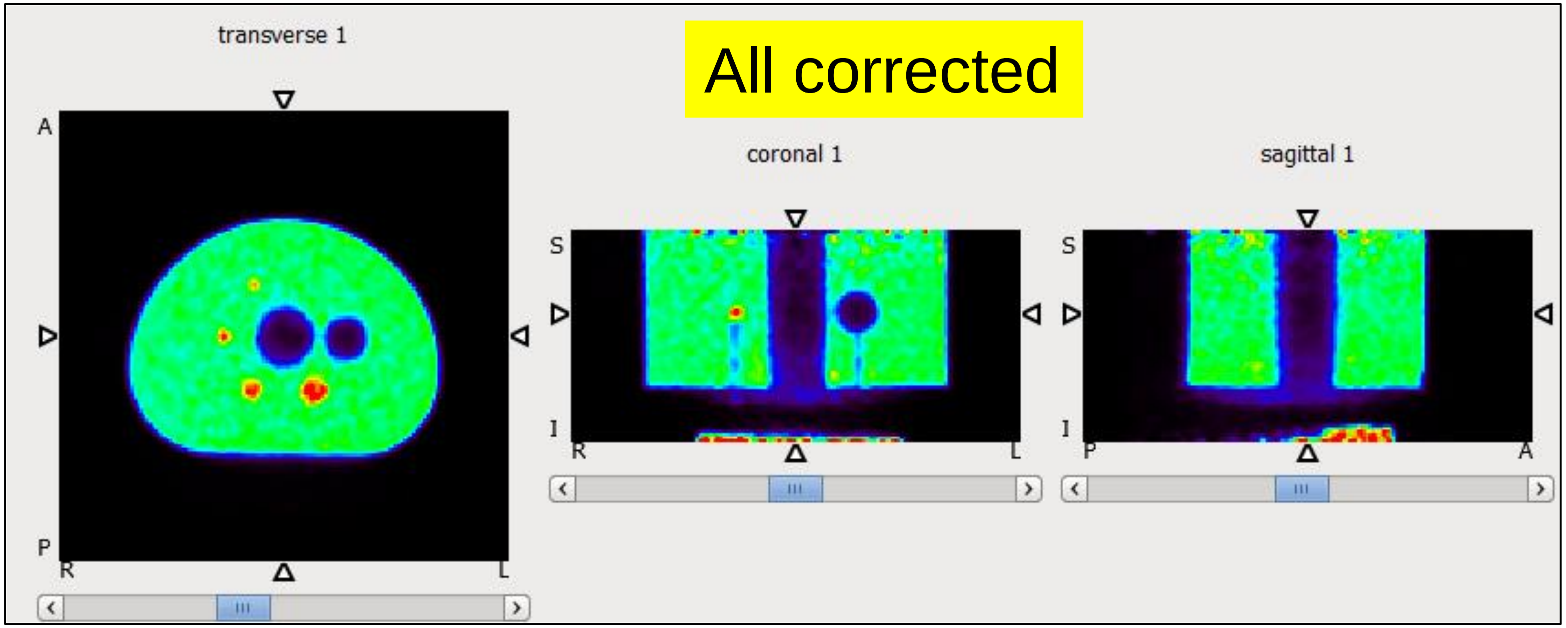
Original CT
512x512

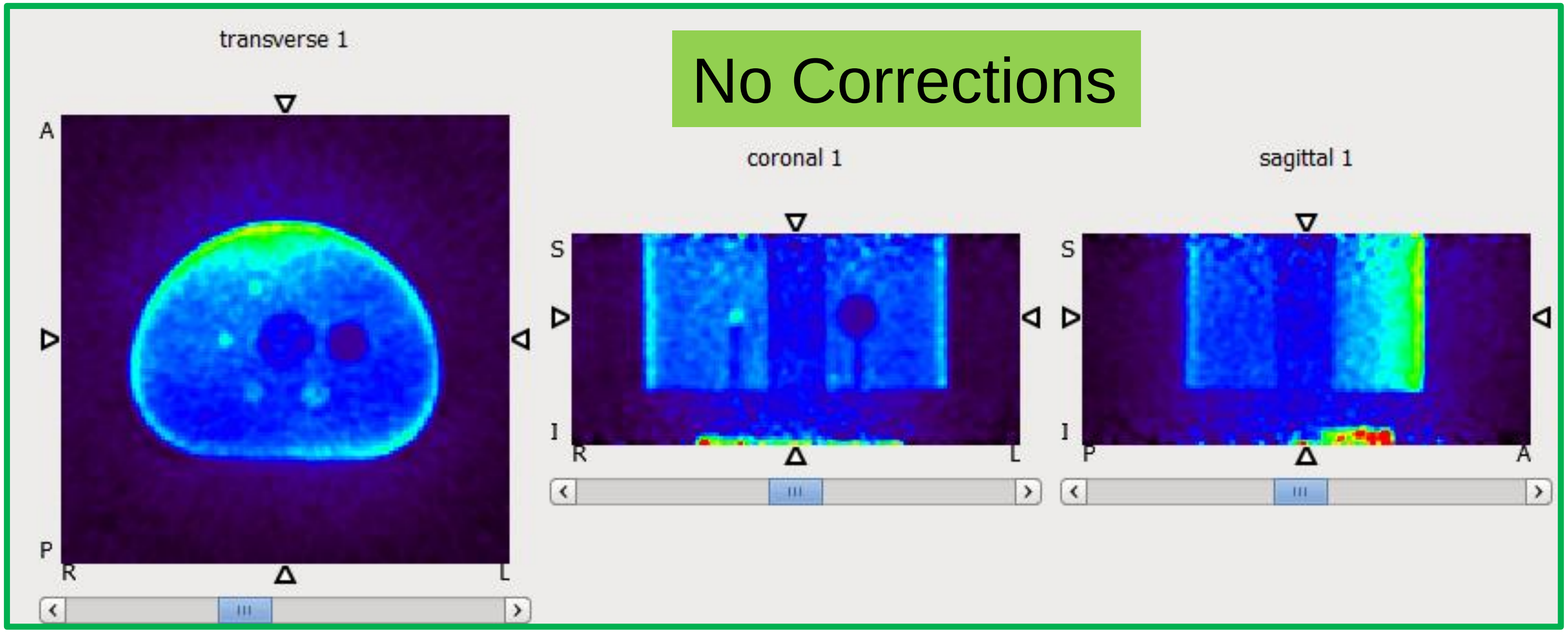
Intermediate CT
128 x 128

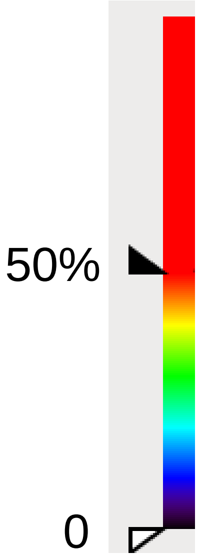
Attenuation
coefficient map
(μ-map)
128 x 128



NEMA Phantom

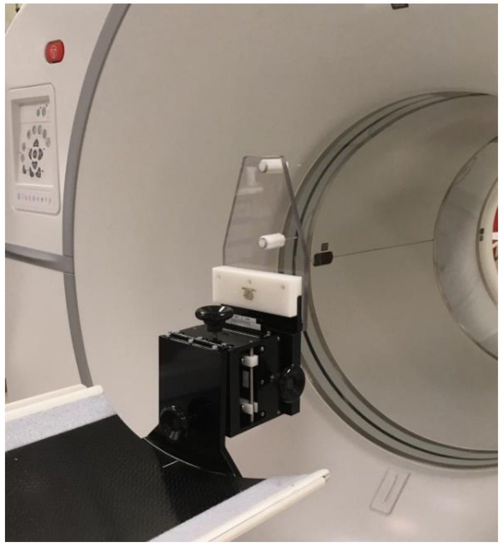




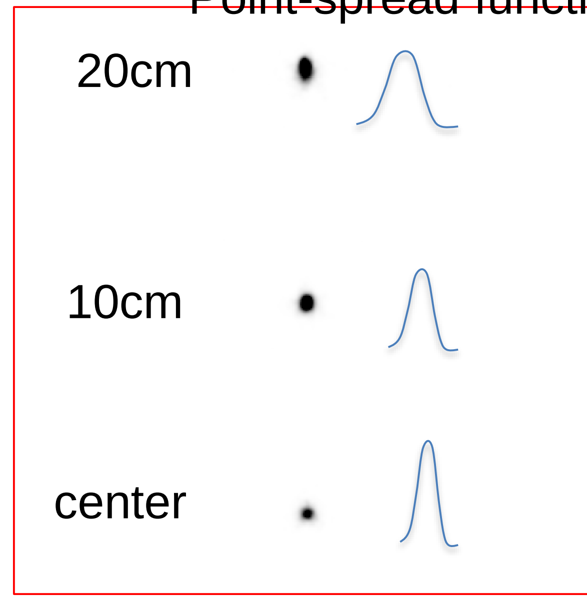


Spatial Resolution modeling in PET imaging

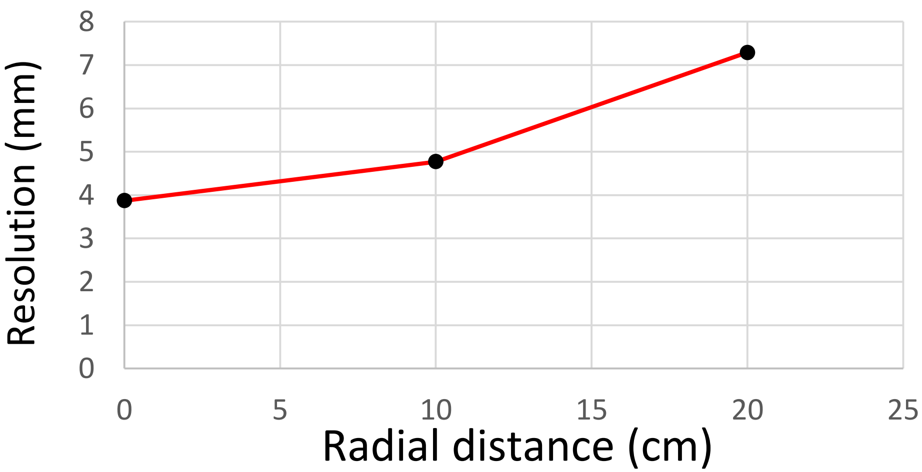
3 line sources at center, 10cm, 20cm radial positions.



Point-spread function



PET image



PSF modeling reconstruction

PET – Better Scintillation detectors

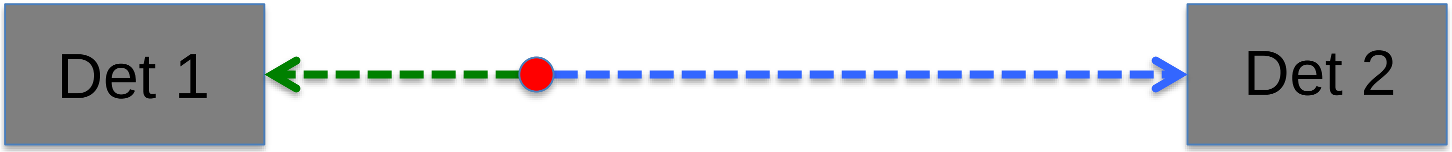
- short decay time of the intensity of light pulse

  300 ns BGO vs. 40ns LYSO

  → to shorten the coincidence timing window (reduce **randoms**)

  → reduce dead-time effect

  → Time of flight imaging

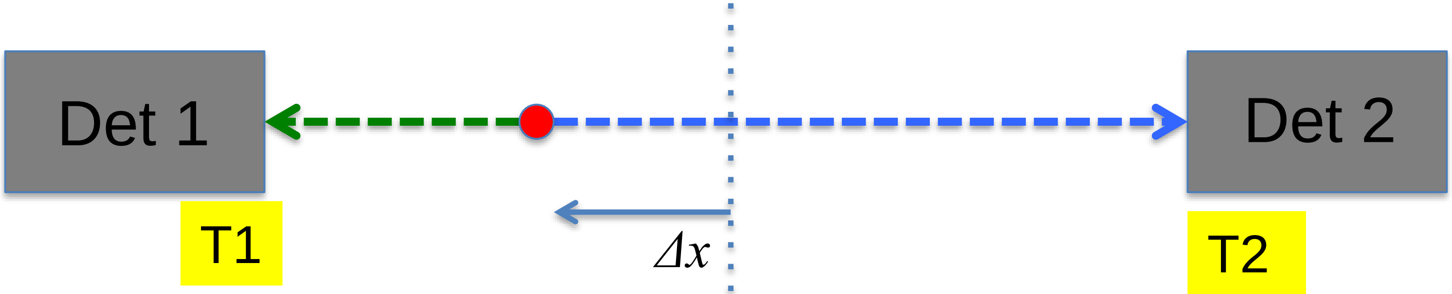


Speed of light (radiation) = 3×10^{10} cm/s = 30,000,000,000 cm/s

Time (ns)	**0.1**	**0.5**	**1.0**	**1.5**	**2**
Distance (cm)	3	15	30	45	60

If Field of view is 50cm, t=d/s = $50/3 \times 10^{10}$ =1.7×10^{-9} sec = 1.7 ns
If Field of view is 20cm, t=d/s = $20/3 \times 10^{10}$ =1.7×10^{-9} sec = 0.7 ns

Time of Flight (TOF)



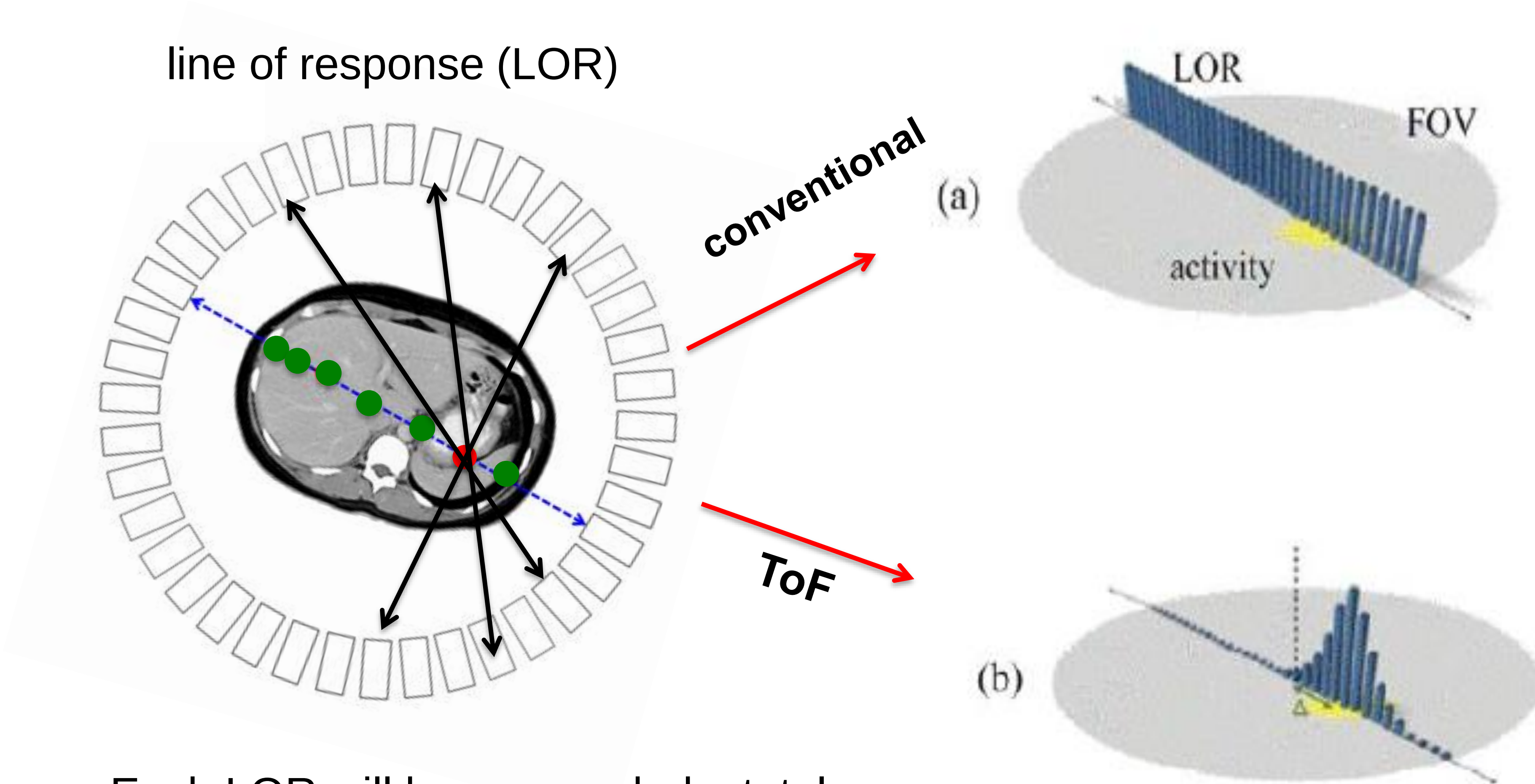
- TOF systems measure the time between each coincidence photon
- determine the event location along the line of response.

$$\Delta x = \frac{\Delta T \times c}{2} \qquad \text{where } \Delta T = T2 - T1$$

ΔT (ns)	0.1	0.2	0.3	0.4	0.5
Δx (cm)	1.5	3.0	4.5	6.0	7.5

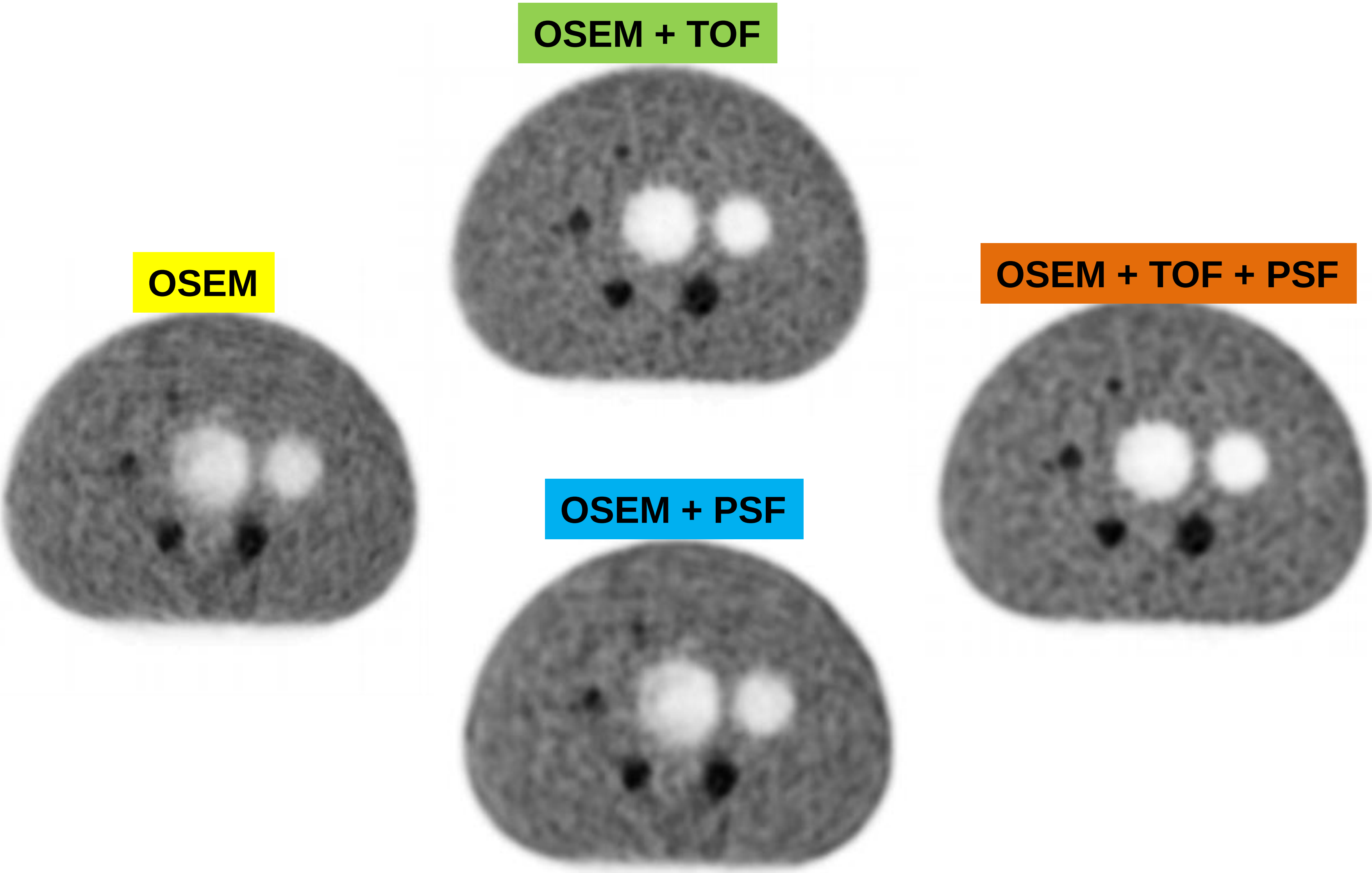
- The event location accuracy can be measured proportionally to the **system's timing resolution**. → need a fast scintillator!
- **LYSO ~ 500 ps (=0.5 ns) , $LaBr_3$ ~ 300 ps (=0.3ns)**

TOF reconstruction



Each LOR will have recorded a total number of counts proportional to the <u>integral</u> of the radioactivity concentration along that LOR.

PET reconstruction and Image Quality



Iterative Ordered Subsets Expectation-Maximization (**OSEM**)

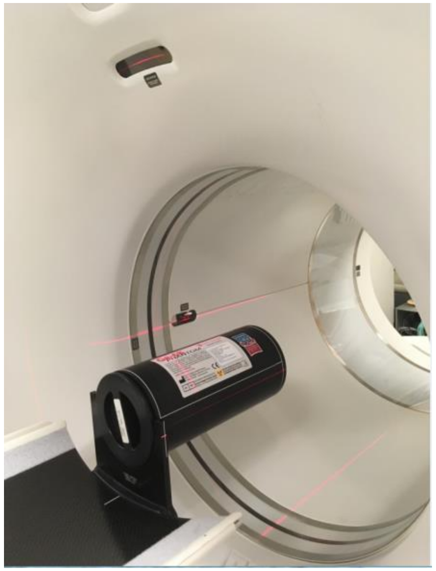
Equipment Quality Control for PET

Detector Normalization

- ✓ similar to flood calibration is gamma camera
- ✓ manufacturer recommended frequency, e.g., quarterly

- ✓ To acquire detector crystal efficiency using
  - a rotating Ge-68 line source or a uniform cylindrical Ge phantom

- ✓ <u>the output of every detector is balanced in order to give the same response as for uniform irradiation.</u>



Blank scan

- ✓ Similar to daily uniform check in SPECT camera
- ✓ To testing for system uniformity

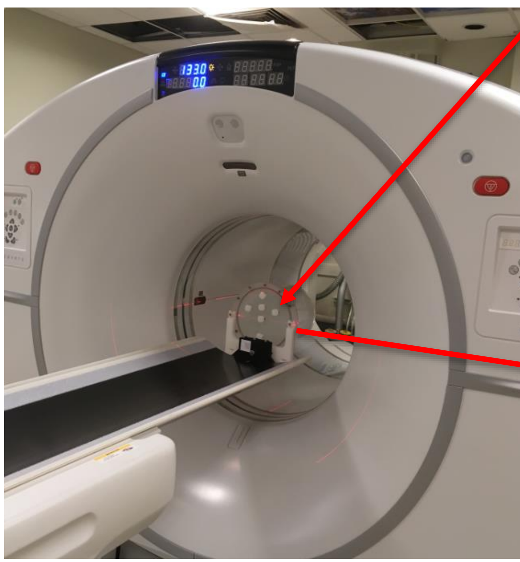
- ✓ Daily test performed prior to patient imaging
- ✓ Use Ge-68 line or cylindrical source for a quick flood imaging

Equipment Quality Control for PET

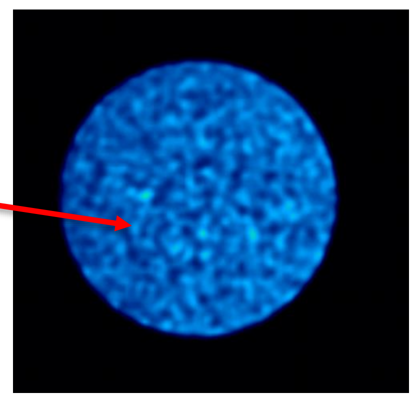
Coincident Timing Calibration

- ✓ Calibration to detect coincident events while reducing random events
- ✓ Using a centrally located source

Dose Calibrator and PET cross calibration, and SUV accuracy



Scan a phantom with
known activity and known volume →
Activity concentration in the phantom is
known



Convert the “**measured count**” into “<u>activity concentration</u>”

Why accredited?

- Under the MIPPA act of 2008, all facilities that bill for **advanced diagnostic imaging** service must be accredited by a CMS designated accrediting organization to quality for Medicare reimbursement for the technical component of those service

  MRI, CT, PET, and Nuclear medicine

- Accreditation is a process in which certification of competency, authority, or credibility is presented.
- Show commitment toward highest standards in the field
- Trustable in the public's eyes

  ✓ The Joint Commission (TJC)
  ✓ Intersectional Accreditation Commission (IAC)
  ✓ The American College of Radiology (ACR)
  ✓ RadSite

Medicare Improvements for Patients and Providers Act of 2008 (H.R. 6331)
CMS : Center for Medicare and Medicaid services

Equipment Quality Control Requirements for PET

In addition to routine tests that are required by manufacturer,

<u>The Joint Commission (& RadSite)</u>

- Image uniformity
- system spatial resolution
- Low-contrast resolution or detectability Artifact evaluation
- Acquisition Monitor test

Optional - sensitivity, energy resolution, and count-rate performance.

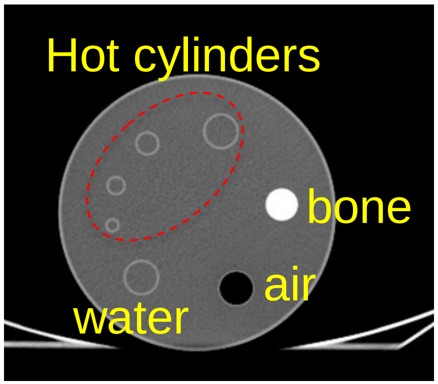
<u>Intersectional Accreditation Commission (IAC)</u>

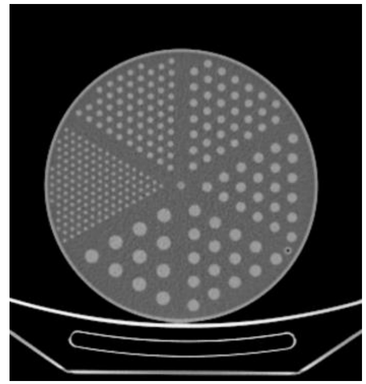
- Daily system test
- Tomographic uniformity
- Normalization to calibrate the efficiency of all detectors
- Absolute activity calibration
- Alignment of the PET and CT scans
- Preventive maintenance

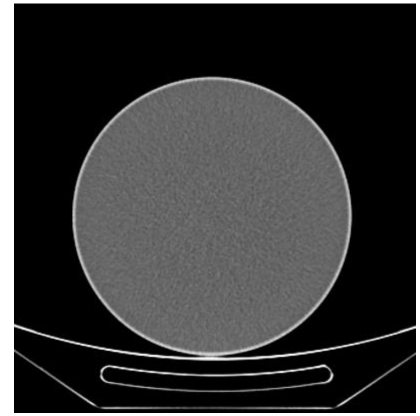
ACR Accreditation

Annual physics testing is required using an ACR approved phantom

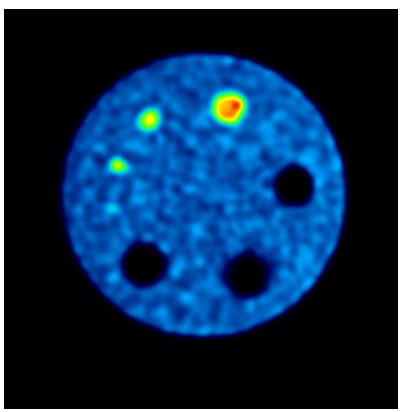
CT

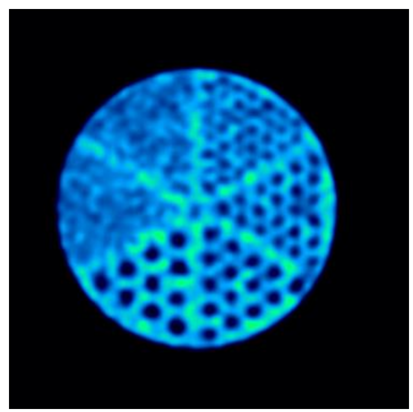


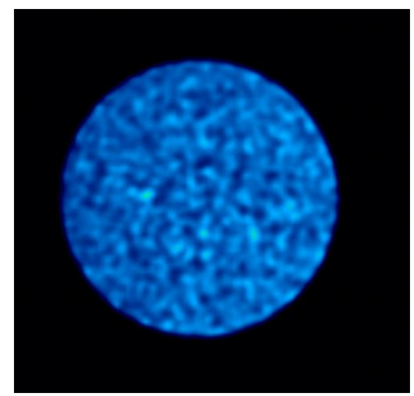




PET







Contrast

Resolution

Uniformity