# Recent Advances in SPECT/CT and PET/CT for Oncology

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# Educational Objectives

- To discuss the physics and describe the recent advances in commercial technology of SPECT/CT and PET/CT for oncology

# SPECT and PET

- Single Photon Emission Computed Tomography
- Positron Emission Tomography

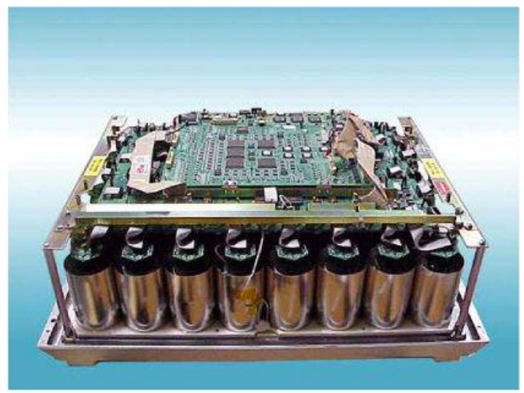
  - Radio-pharmaceutical administration – injected, ingested, or inhaled
  - Bio-distribution of pharmaceutical – uptake time
  - Decay of radionuclide from within the patient – the source of information
  - SPECT – Gamma camera detects radionuclide emission photons
  - PET – Coincidence ring detector detects annihilation photons
  - Tomography performed to image the radio-pharmaceutical distribution within the patient

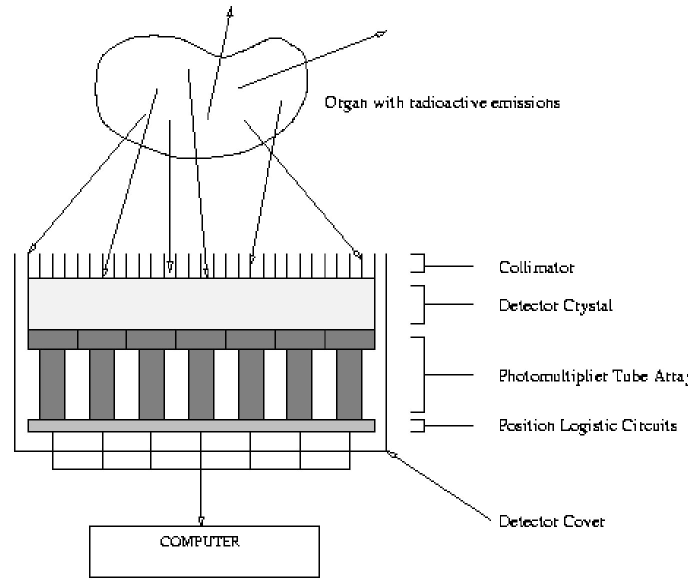
- Used for visualization of functional information based on the specific radio-pharmaceutical uptake mechanism

# SPECT/CT

# Gamma Camera

- NaI(Tl) is the scintillator of choice
  - High light output and High detection efficiency (~85% at 140 keV for 3/8 in. NaI)
  - Good energy resolution (~10% at 140 keV)
  - Large crystals (50 cm x 40 cm)
  - Hygroscopic!

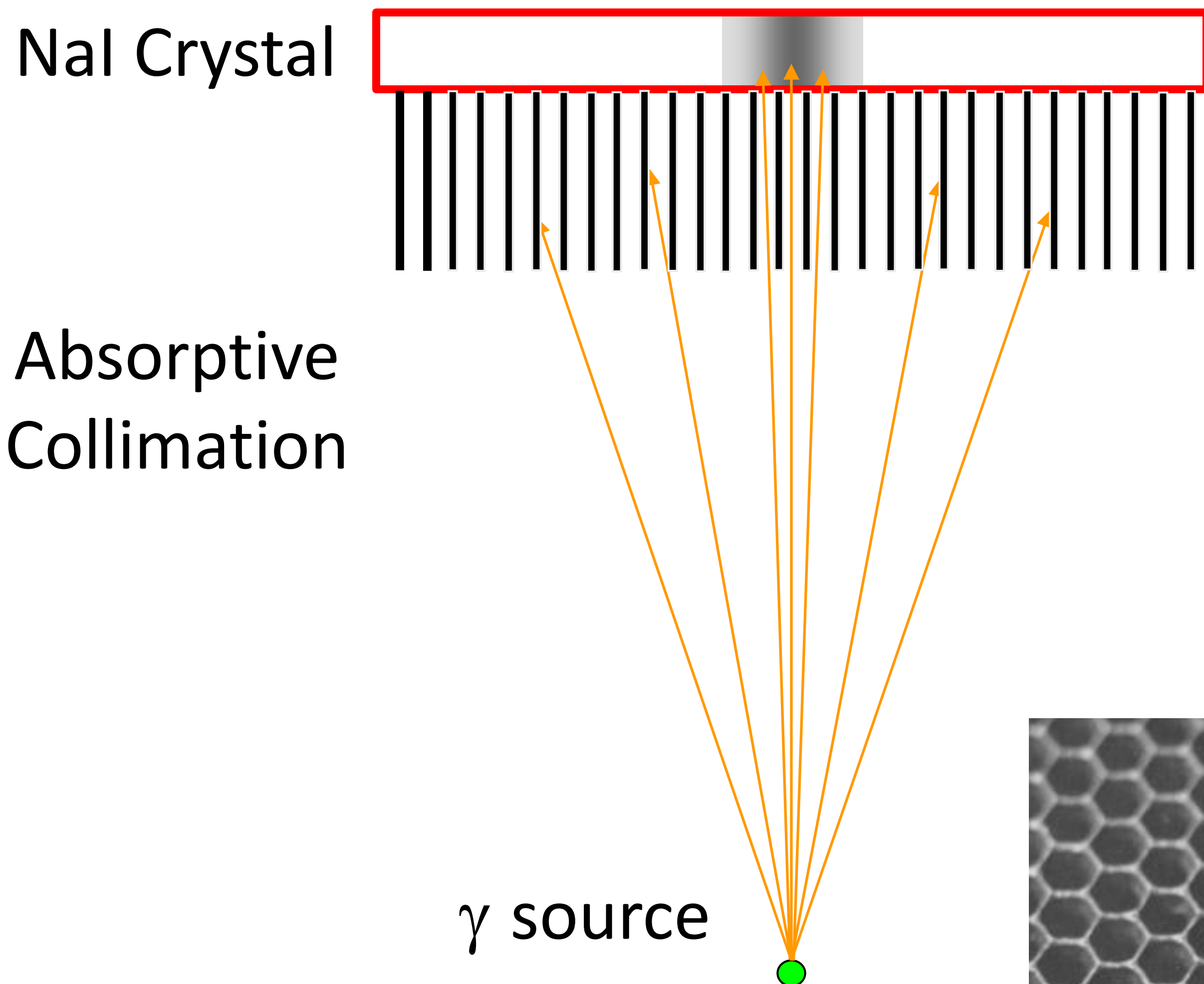


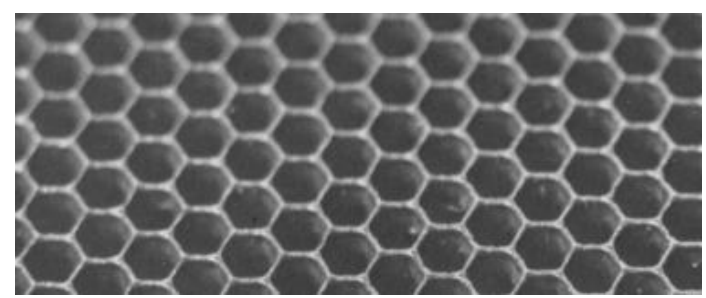




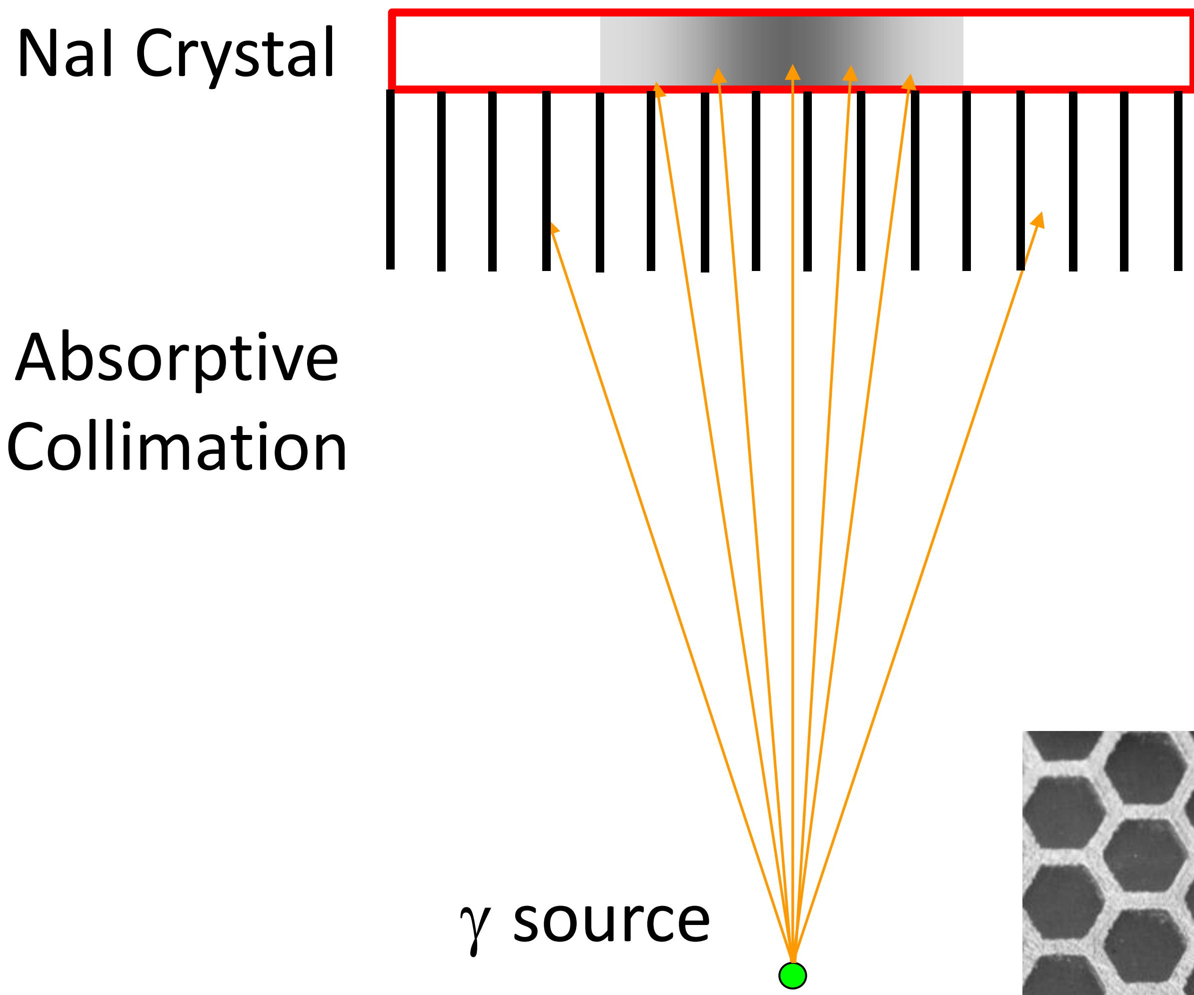

- Intrinsic Spatial and Energy Resolution
  - # of scintillation photons, $N \propto$ Gamma-ray energy, $E$
  - Spatial Resolution = $100 \times \sigma/N \propto 1/\sqrt{N} \propto 1/\sqrt{E}$
  - Energy Resolution = $100 \times FWHM/E \propto 1/\sqrt{E}$

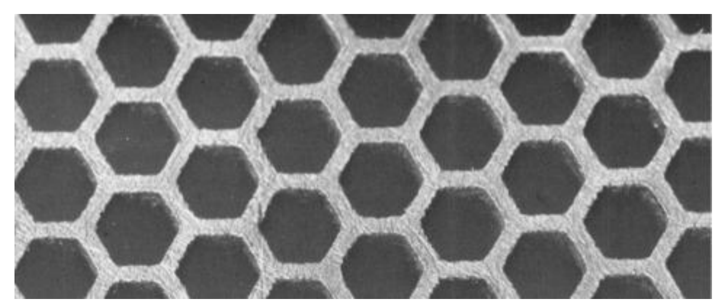
# Collimators



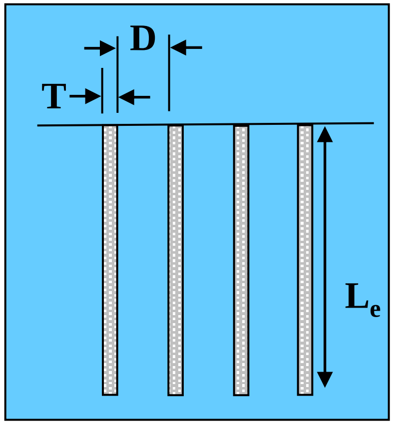



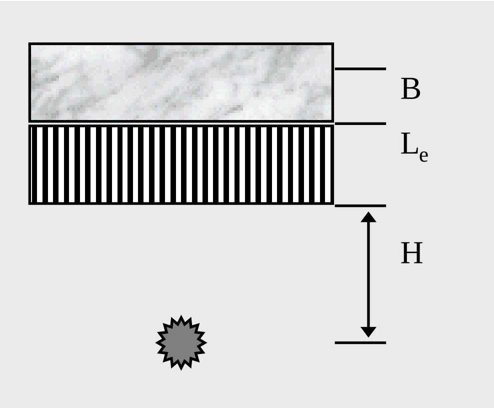
# Collimators

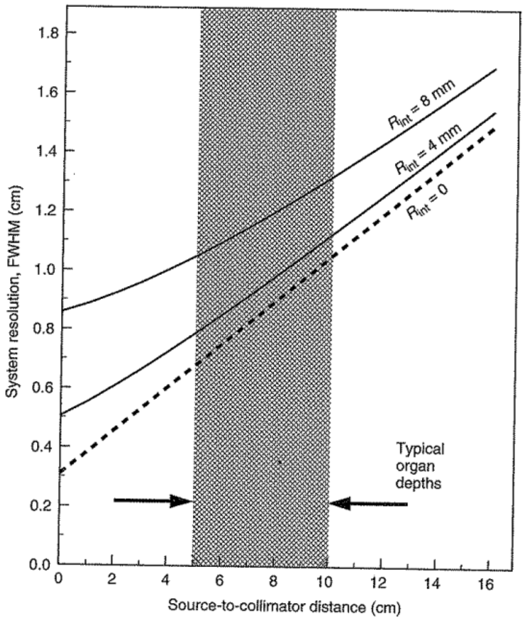


# Collimator Resolution




*Cherry, Sorenson, & Phelps, Physics of Nuclear Medicine, 2003*

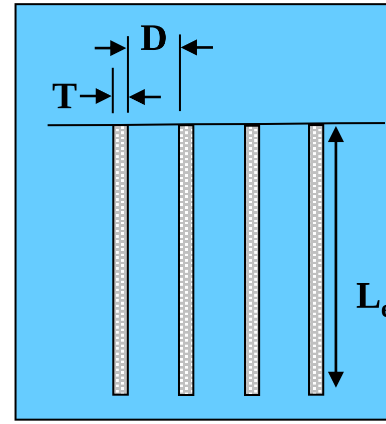
Collimator Resolution $$R_g = \frac{D(L_e + H + B)}{L_e}$$

System Resolution $$R_s^2 = R_i^2 + R_g^2$$

# Collimator Efficiency

$$G = \theta F \text{ where } \theta = C(D/L_e)^2$$

$$G = \frac{CD^4}{L_e^2(D+T)^2}$$




θ = fraction of 4π

F = exposed fraction

Parallel Hexagonal hole C = 3/8π

LEHR = $1.3\times10^{-4}$

MELP = $3.1\times10^{-4}$

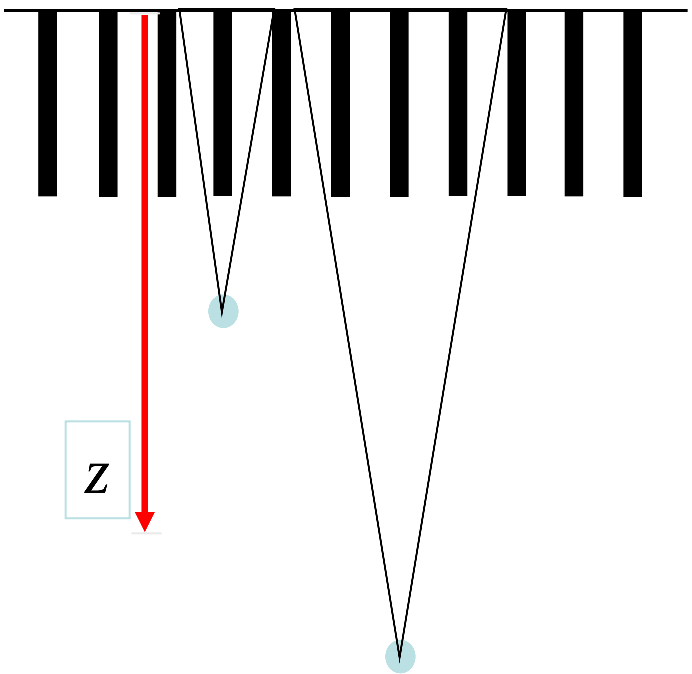
**BiCoreTM Collimators Specifications**

| Collimators | LEHS Low Energy High Sensitivity | LEAP Low Energy All Purpose | LEHR Low Energy High Resolution | LEUHR Low Energy Ultra High Resolution | LEFB Low Energy Fan Beam | MELP Medium Energy Low Penetration | HE High Energy | UHE Ultra High Energy |
|---|---|---|---|---|---|---|---|---|
| Isotope | $^{99m}$Tc | $^{99m}$Tc | $^{99m}$Tc | $^{99m}$Tc | $^{99m}$Tc | $^{67}$Ga | $^{131}$I | $^{18}$F |
| Hole Shape | Hex | Hex | Hex | Hex | Hex | Hex | Hex | Hex |
| Number of Holes (x1000) | 28 | 90 | 148 | 146 | 64 | 14 | 8 | 4 |
| Hole Length (mm) | 24.05 | 24.05 | 24.05 | 35.8 | 35 | 40.64 | 59.7 | 50.5 |
| Septal Thickness (mm) | 0.36 | 0.2 | 0.16 | 0.13 | 0.16 | 1.14 | 2 | 3.4 |
| Hole Diameter (mm across the flats) | 2.54 | 1.45 | 1.11 | 1.16 | 1.53 | 2.94 | 4 | 2.5 |
| Sensitivity @ 10 cm1 (cpm/_Ci) | 1020 | 330 | 202 | 100 | 280 | 310 | 147 | 185 |
| Geometric Resolution @ 10 cm (mm) | 14.6 | 8.3 | 6.4 | 4.6 | 6.3 | 10.8 | 13.2 | 10.6 |
| System Resolution @ 10 cm1 (mm) | 15.6 | 9.4 | 7.4 | 6.0 | 7.3 | 12.5 | 13.4 | 19.0 |
| Septal Penetration (%) | 1.5 | 1.9 | 1.5 | 0.8 | 1.0 | 1.2 | 3.5 | 3.4 |
| Focal Length @ Exit Surface (mm) | n.a. | n.a. | n.a. | n.a. | 445 | n.a. | n.a. | n.a. |
| Weight (lb) | 42 | 49 | 45 | 56 | 67 | 136 | 296 | 260 |
| Weight (kg) | 18.9 | 22.1 | 20.4 | 25.2 | 30.5 | 61.8 | 134.5 | 117.0 |

1. Values measured in accordance with NEMA Standards Publication NU-1 2001 using 3/8" crystal.

# Sensitivity versus Source Distance

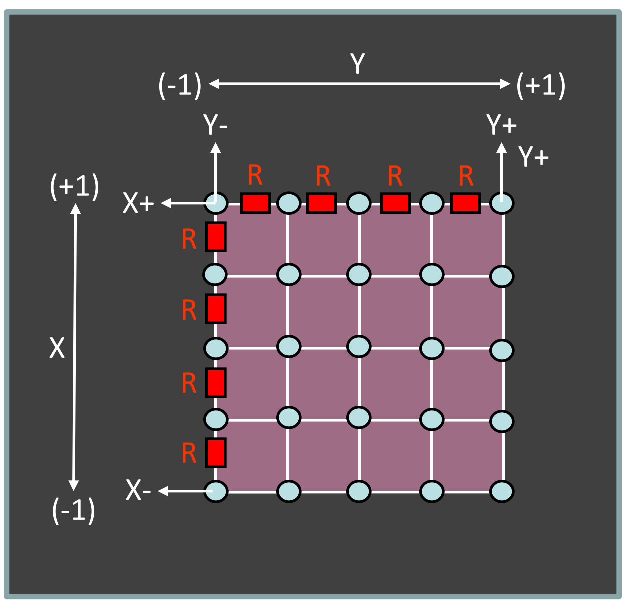
- Sensitivity: the detected photons count rate per unit activity [cps/uCi]
- Photon flux vs. distance
  $\propto z^{-2}$
- Crystal area vs. distance
  $\propto z^{2}$
- Overall sensitivity
  $S \propto z^{-2} \times z^{2} \sim$ constant

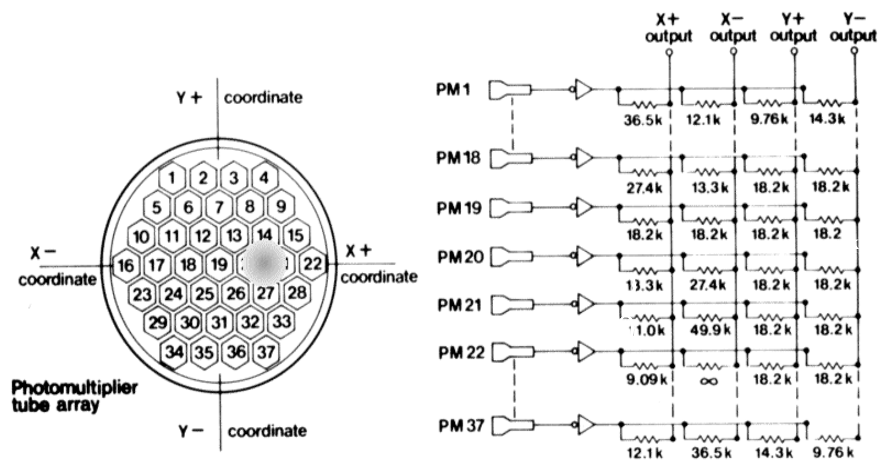



$$z = L_e + H + B$$

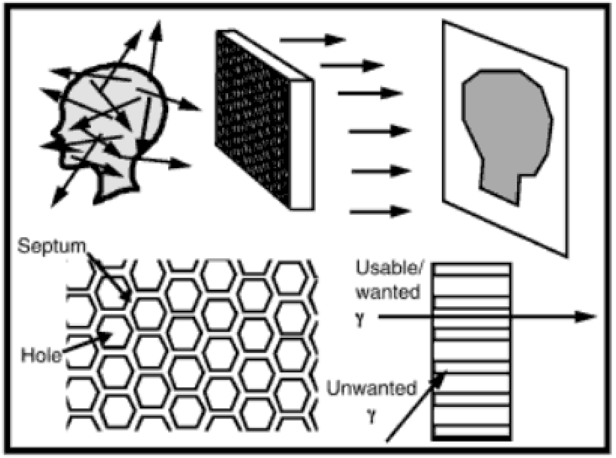
# Anger Logic for Event Position

- Interaction location based on relative signal between $X^+$ and $X^-$ (for X location) & $Y^+$ and $Y^-$ (for Y location)
  - $X = (X^+ - X^-)/(X^+ + X^-)$ → range -1 to +1
  - $Y = (Y^+ - Y^-)/(Y^+ + Y^-)$ → range -1 to +1
- Interaction Energy ∝ Total Signal = $X^+ + X^- + Y^+ + Y^-$

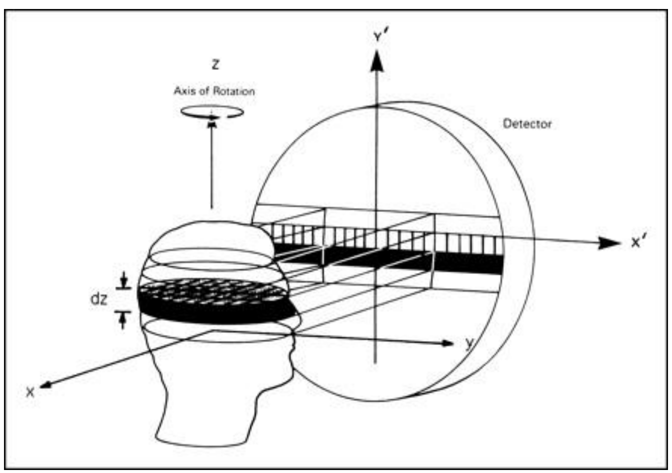


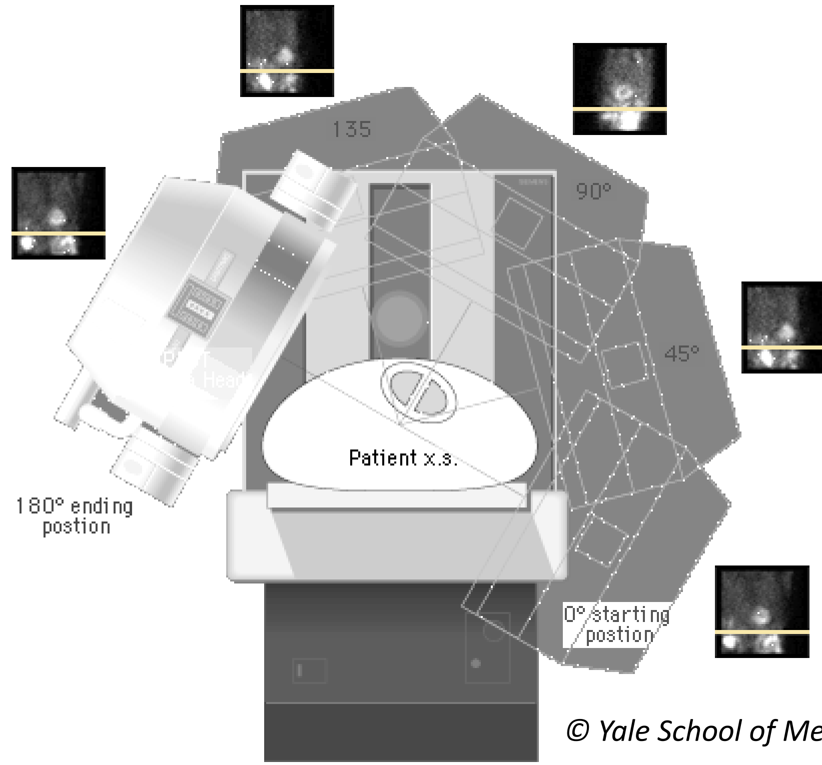
# SPECT Acquisitions




*Wernick & Aarsvold, Emission Tomography, 2004*

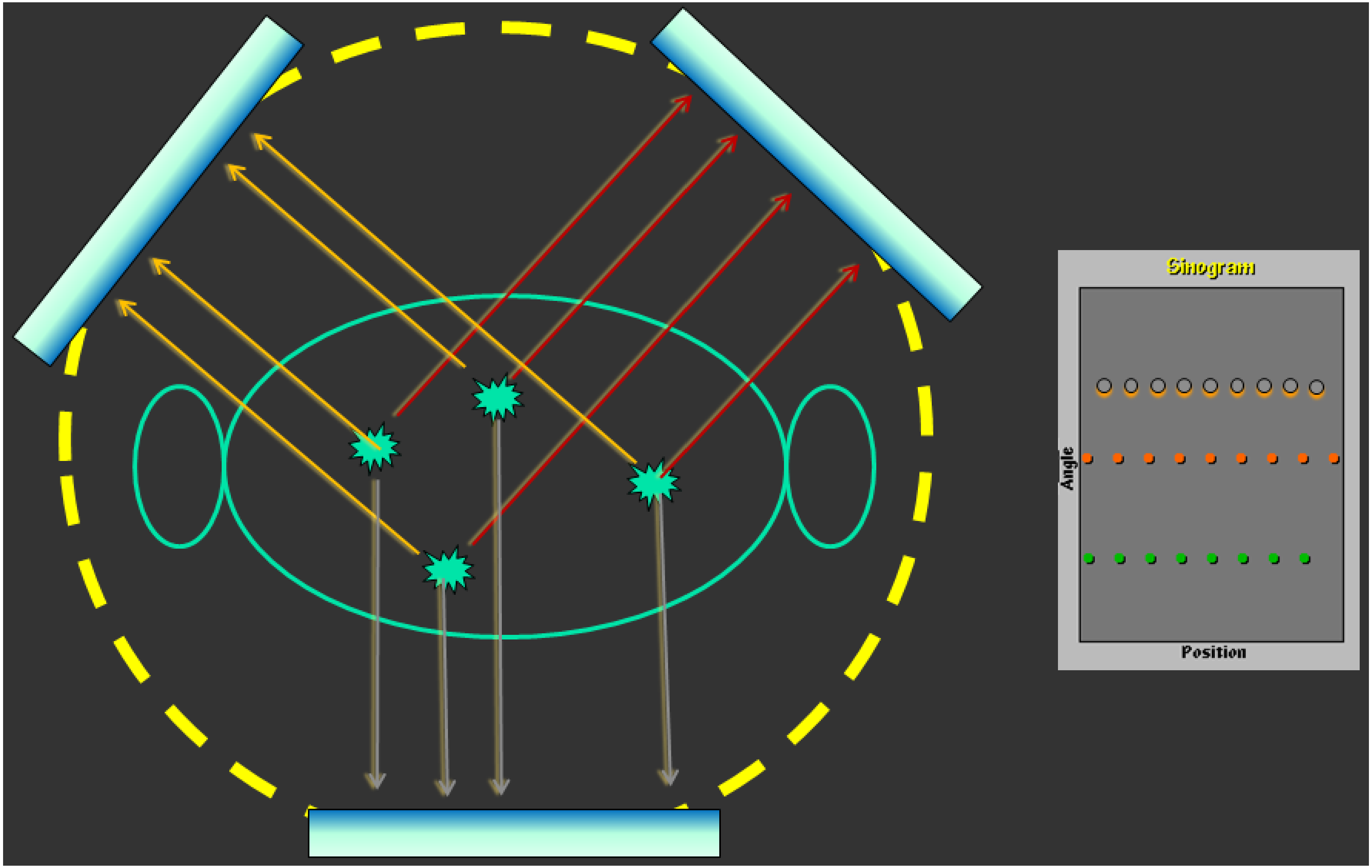


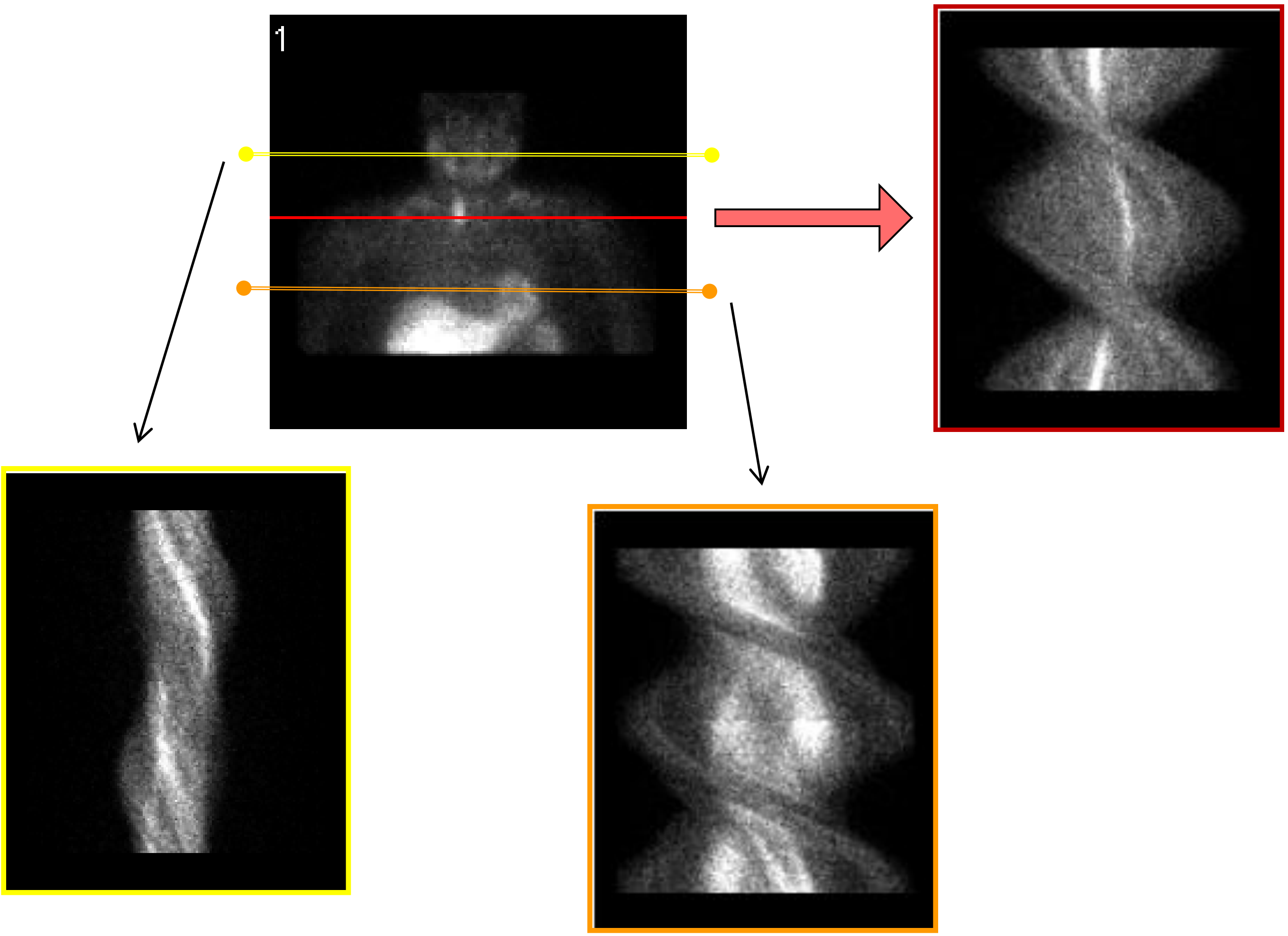

*SPECT in the year 2000, JNMT 24:233, 2000*



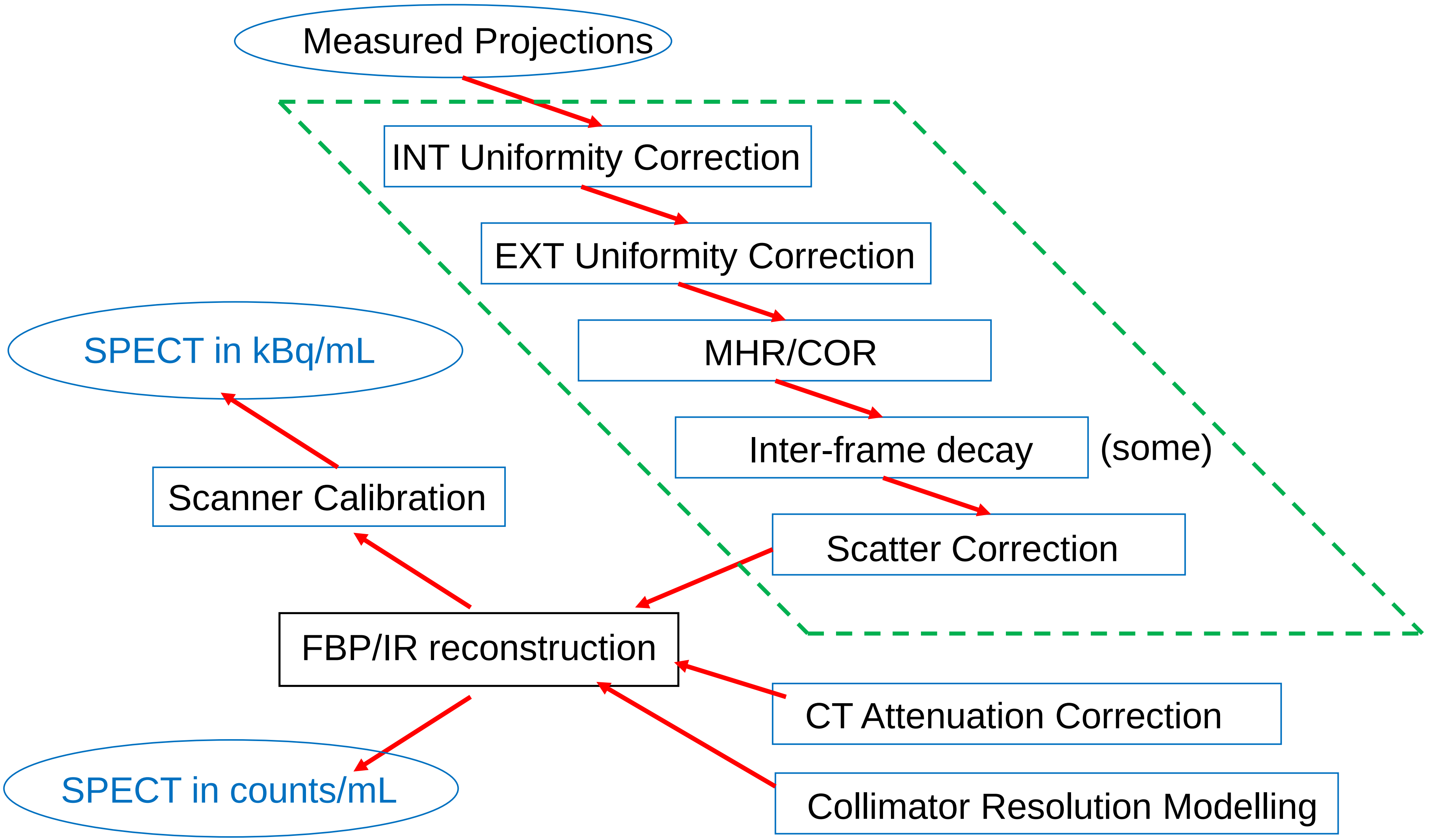

*© Yale School of Medicine*

SPECT acquires 2D projections of a 3D volume


Sinogram
Angle
Position


1

# SPECT data corrections



# SPECT Iterative Recon: Scatter Modeling

- Scatter compensation occurs before attenuation
  - the photopeak window contains scatter
  - attenuation accounts for the removal of photopeak photons

- Adjacent energy window based estimate (DEW and TEW): Scatter estimated as a weighted sum of adjacent energy window images, $Ci(x,y,\theta)$

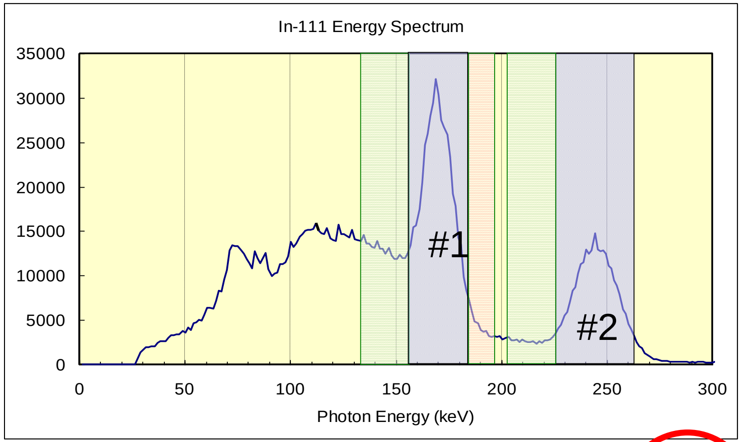
$$S(x,y,\theta) = \Sigma_i\, k_i \times C_i(x,y,\theta)$$

- Subtract scatter prior to reconstruction

$$P_{corr}(x,y,\theta) \rightarrow P(x,y,\theta) - S(x,y,\theta)$$

- Incorporate scatter into forward projection

$$P(x,y,\theta) \rightarrow P_{corr}(x,y,\theta) + S(x,y,\theta)$$




TEW

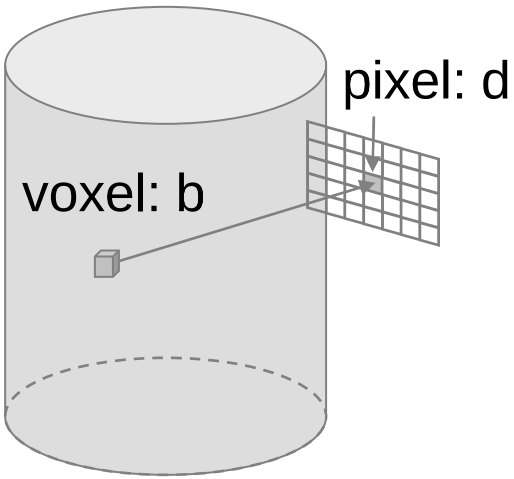
DEW

| Energy Window Definition | | Energy Range | SF |
|---|---|---|---|
| Lower Scatter | 15% | 131 - 156 | 0.5 |
| Photopeak #1 | 172 keV w/ 15% | 156 - 181 | - |
| Upper Scatter | 8% | 181 - 195 | 0.9375 |
| | | | |
| Lower Scatter | 10% | 201 - 225 | 0.75 |
| Photopeak #2 | 247 keV w/ 15% | 225 - 262 | - |

# SPECT Iterative Reconstruction

## Maximum Likelihood Expectation Maximization (ML-EM)
## Ordered Subset Expectation Maximization (OS-EM)

- Accounts for the statistical nature of photon detection
- Incorporates the system response p(b,d) – the probability that a photon emitted from an object voxel *b* is detected by projection pixel *d*
- p(b,d) captures…
  1. Depth-dependent resolution
  2. Position-dependent scatter
  3. Depth-dependent attenuation




$$a_{i,j,k} = a^{\mathrm{AC}}_{i,j,k} \times a^{\mathrm{collimator}}_{i,j,k} \times a^{\mathrm{efficiency}}_{i}$$

- Use a measured attenuation map along with models of scatter and camera resolution to perform a far more accurate reconstruction

# SPECT Iterative Reconstruction

- True projection intensity = sum of true voxel intensities weighted by detection probabilities

Forward Projection

$$y(d) = \sum_{b=1}^{B} \lambda(b)\, p(b,d)$$

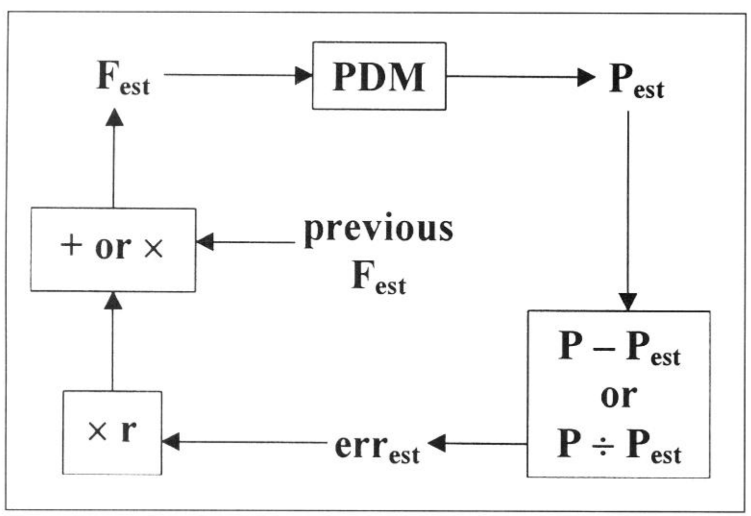
- True voxel intensity = sum of true detector intensities weighted by detection probabilities

Back Projection

$$\lambda(b) = \sum_{d=1}^{D} y(d)\, p(b,d)$$

# Iterative Reconstruction Flow Diagram

$$\lambda^{[k+1]}(b) = \frac{\lambda^{[k]}(b)\sum_{d=1}^{D}\frac{y(d)\,p(b,d)}{\sum_{b'=1}^{B}\lambda^{[k]}(b')\,p(b',d)}}{\sum_{d=1}^{D}p(b,d)}$$




In clinical practice, the stopping criteria is number of iterations (a time constraint) instead of a convergence criteria.

# SPECT Reconstructions

IT=1

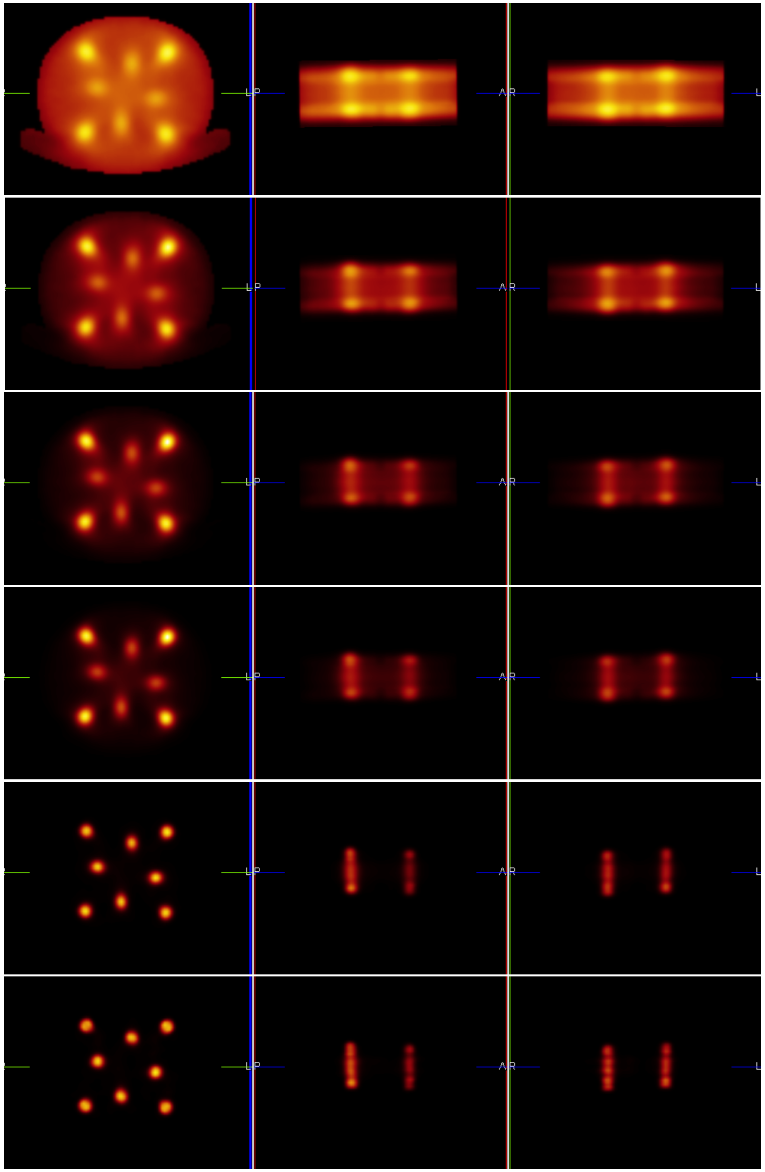
IT=2

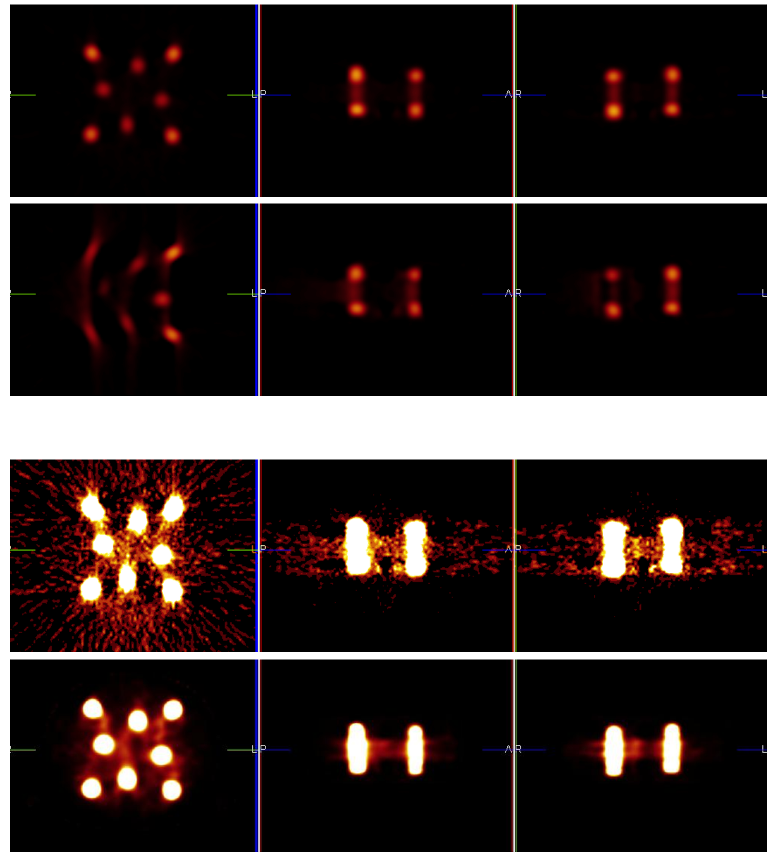
IT=3

IT=4

IT=16

IT=64





FBP $360^0$

FBP $180^0$

FBP noise

IR noise

# HU-to-μ (CT-AC) Transforms

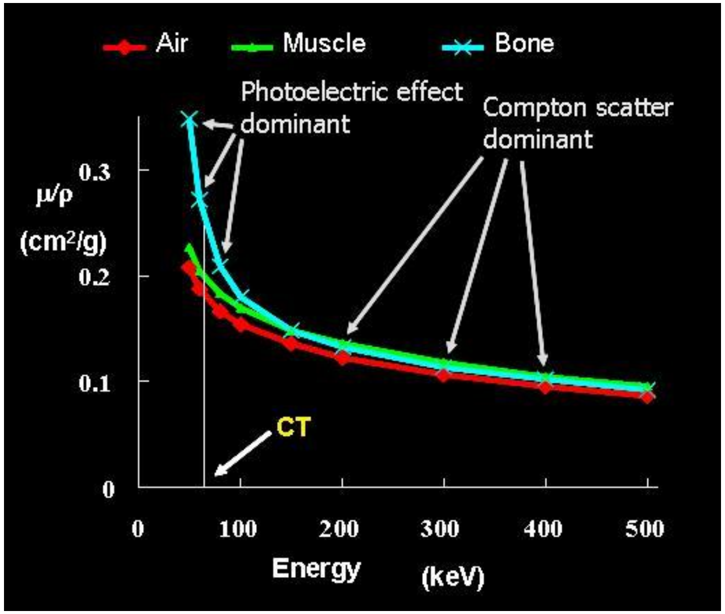
*LaCroix et al., IEEE TNS 41, 1994*

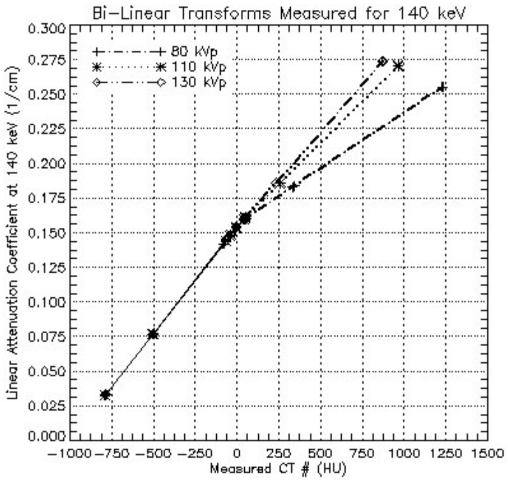
$$HU_x = \frac{\mu_x(E_{CT}) - \mu_w(E_{CT})}{\mu_w(E_{CT})} \times 1000$$

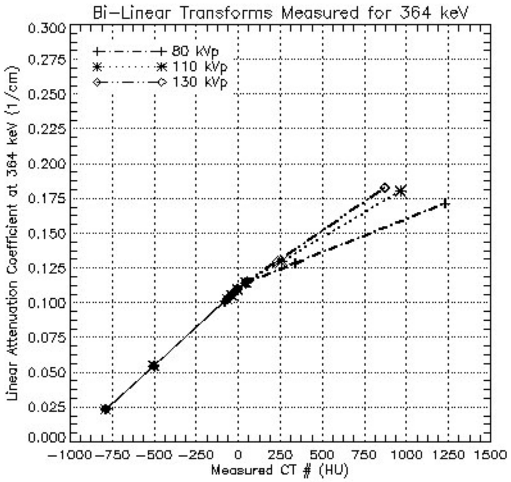
$$\mu_x(E_{CT}) = \left(1 + \frac{HU_x}{1000}\right) \times \mu_w(E_{CT})$$

$$\mu_x(E) = \left(1 + \frac{HU_x}{1000}\right) \times \mu_w(E) \times \underbrace{\left(\frac{\mu_w(E_{CT})}{\mu_x(E_{CT})} \times \frac{\mu_x(E)}{\mu_w(E)}\right)}_{K}$$

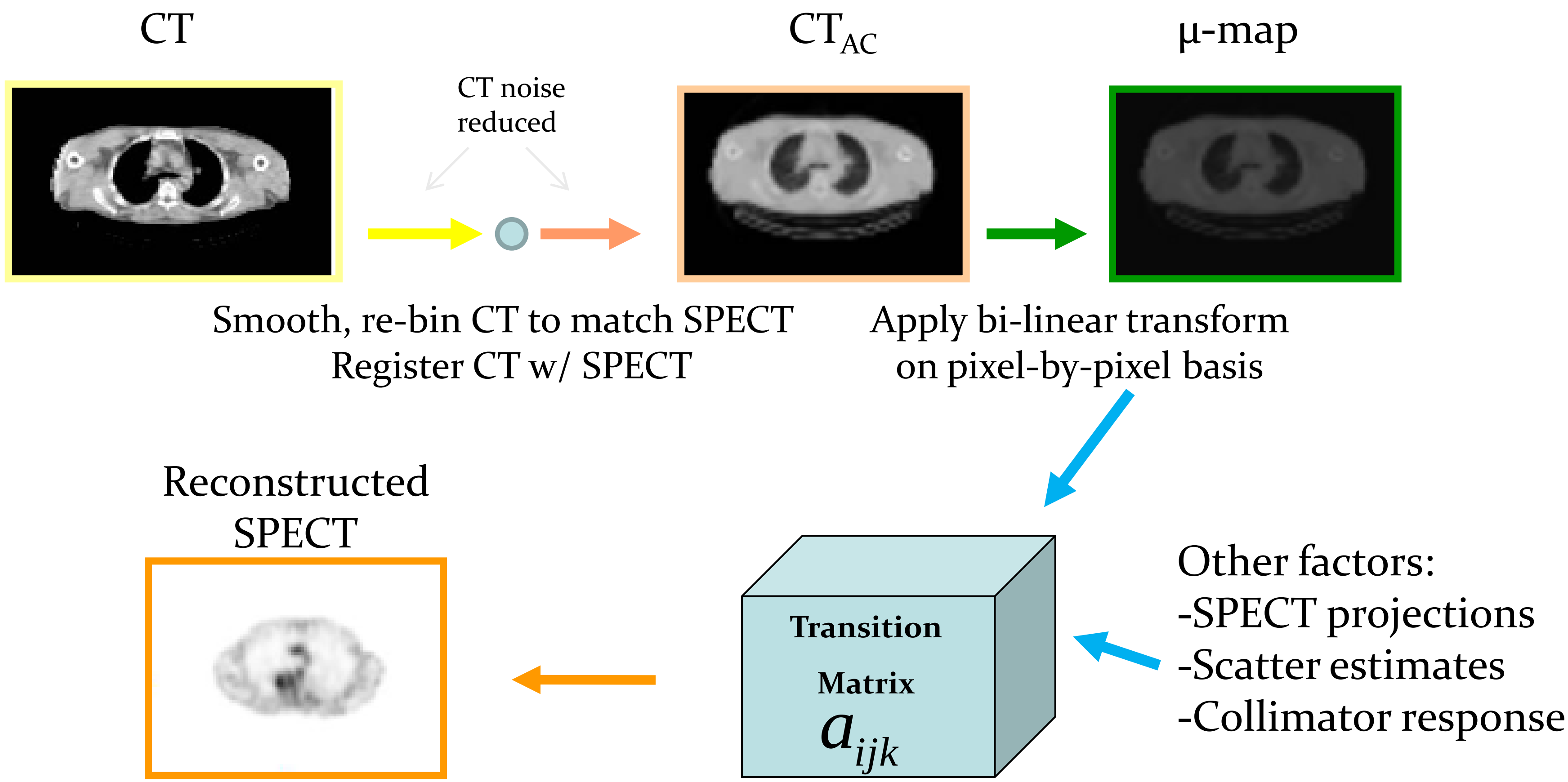
- Photon energies different between CT and SPECT
- K≈1 for Compton Scatter dominates low Z at ECT (low HU)
- K≠1 for Photoelectric pertinent for high Z at ECT (high HU)
- HU-to-μ transform is piece-wise linear (bi- or tri-modal)



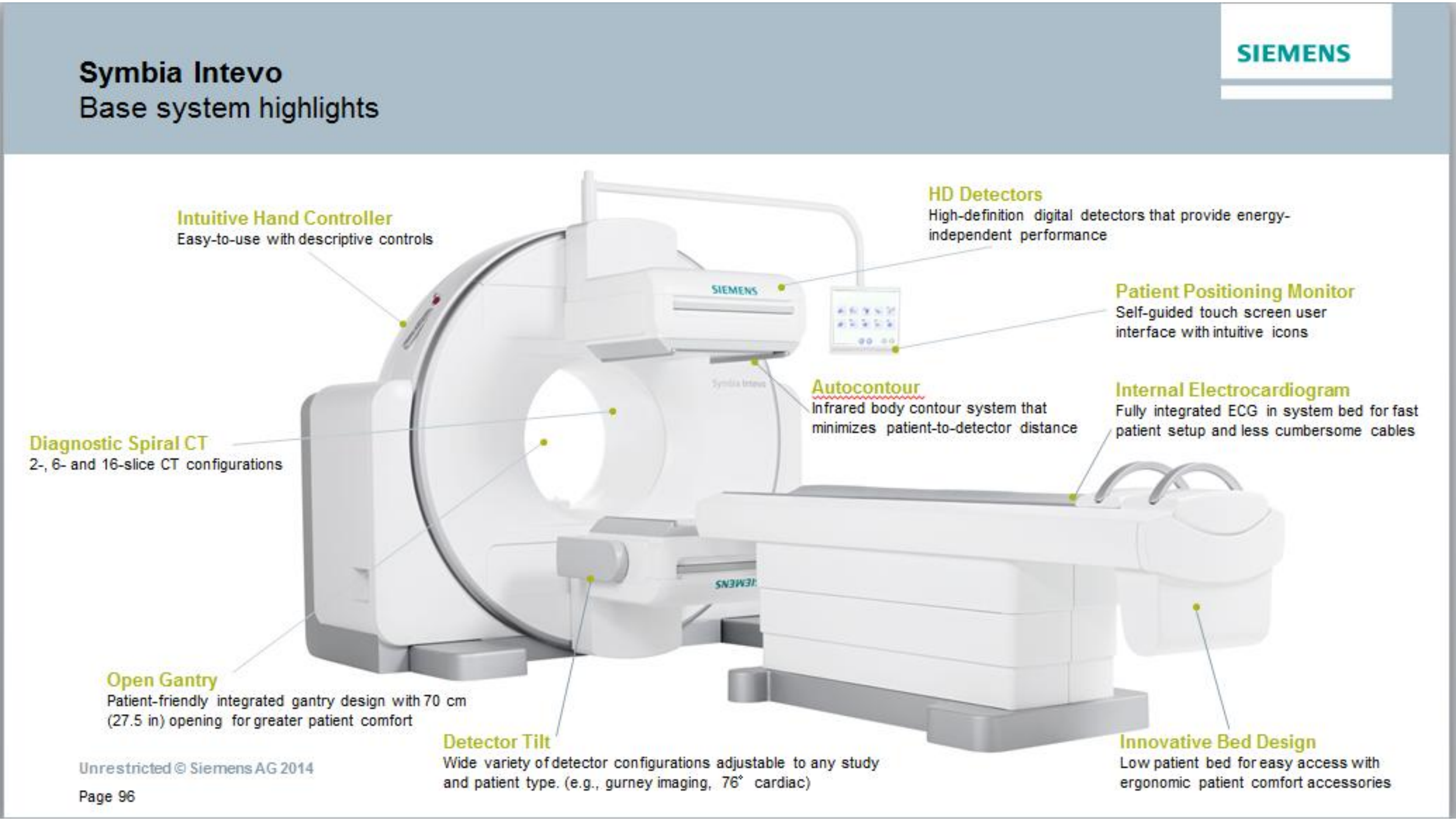



# CT-based AC for SPECT/CT



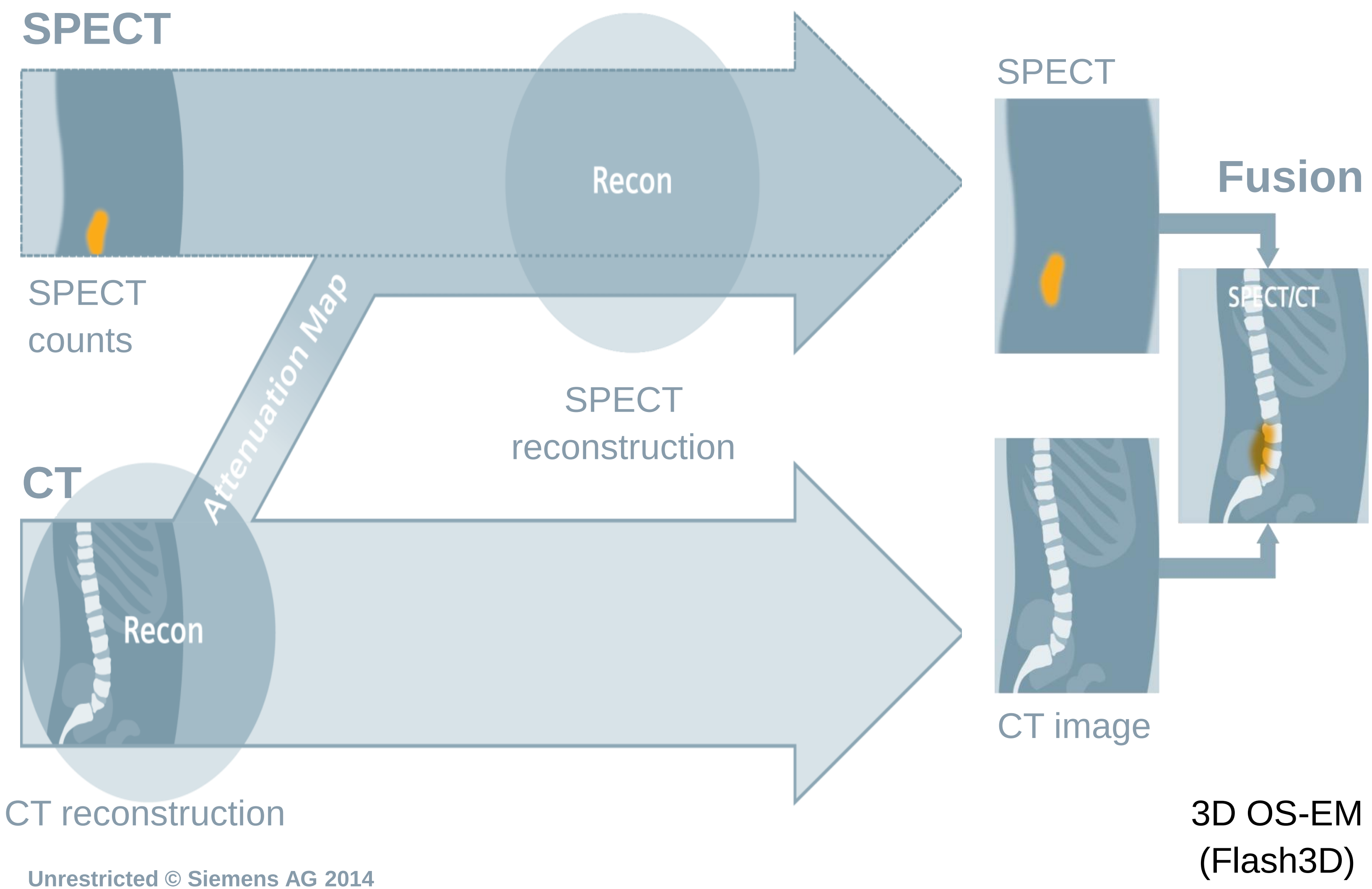
# Siemens – Symbia Intevo



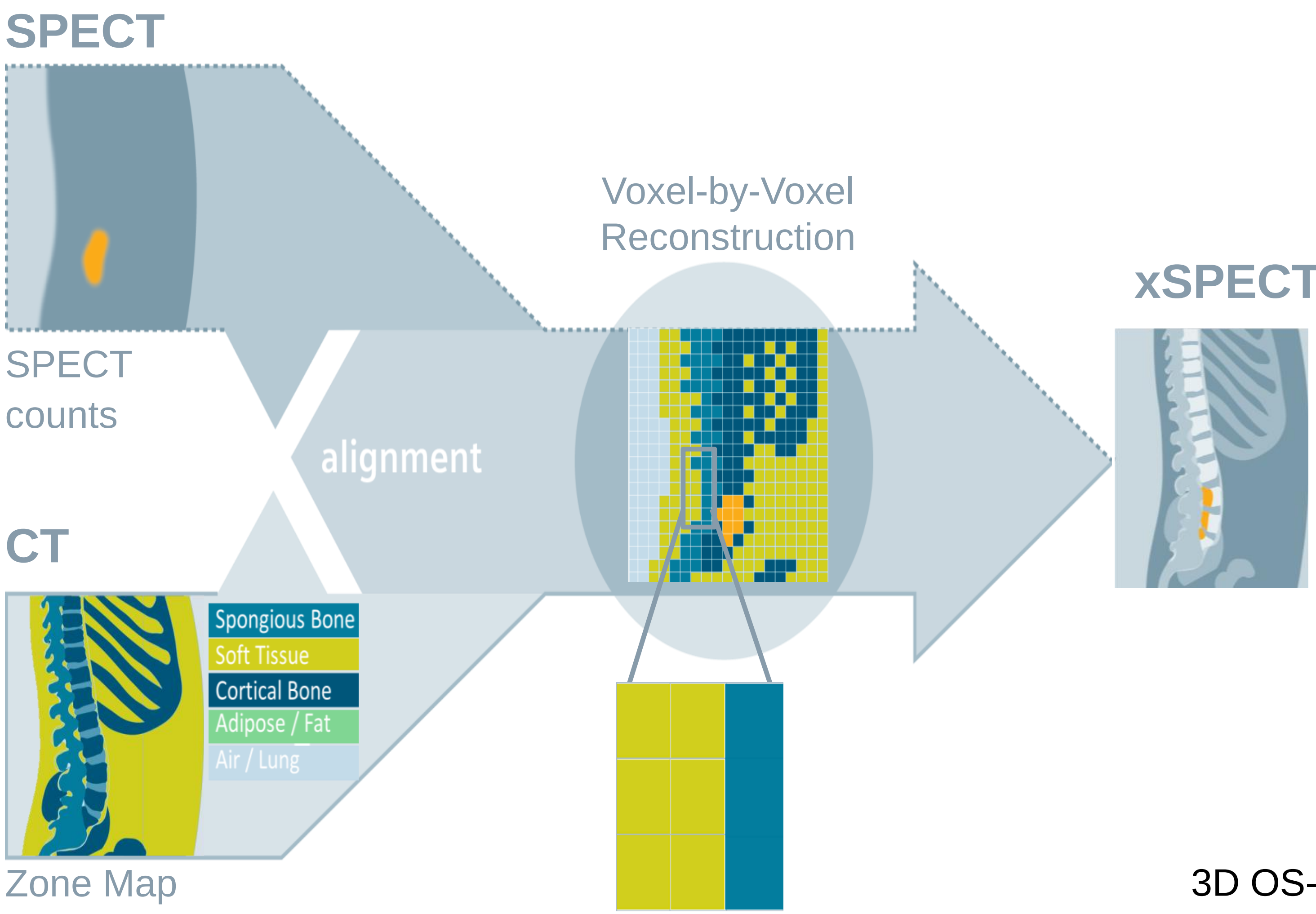

Diagnostic CT
Quantitative SPECT
Advanced SPECT/CT reconstruction

# Conventional SPECT/CT Technology

Mechanical fusion of SPECT and CT



# See the Unseen
Differentiation of tissue boundaries in bone imaging




3D OS-CGM
w/ CT-based Zones

SIEMENS

## See the Unseen

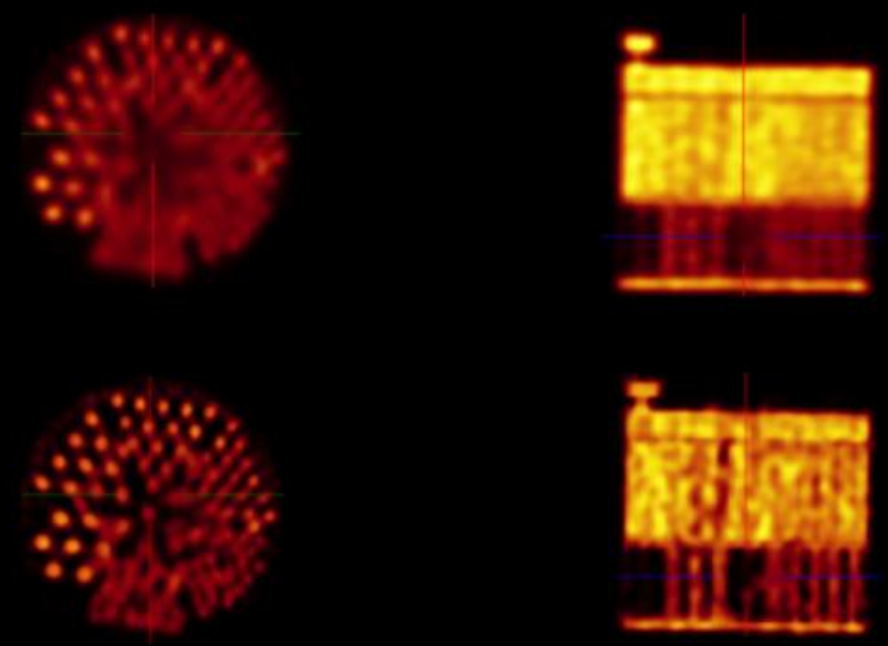
xSPECT reconstruction shows better image quality

**Conventional SPECT**

- OSEM 3D iterative

**xSPECT**

- OSCGM 3D iterative



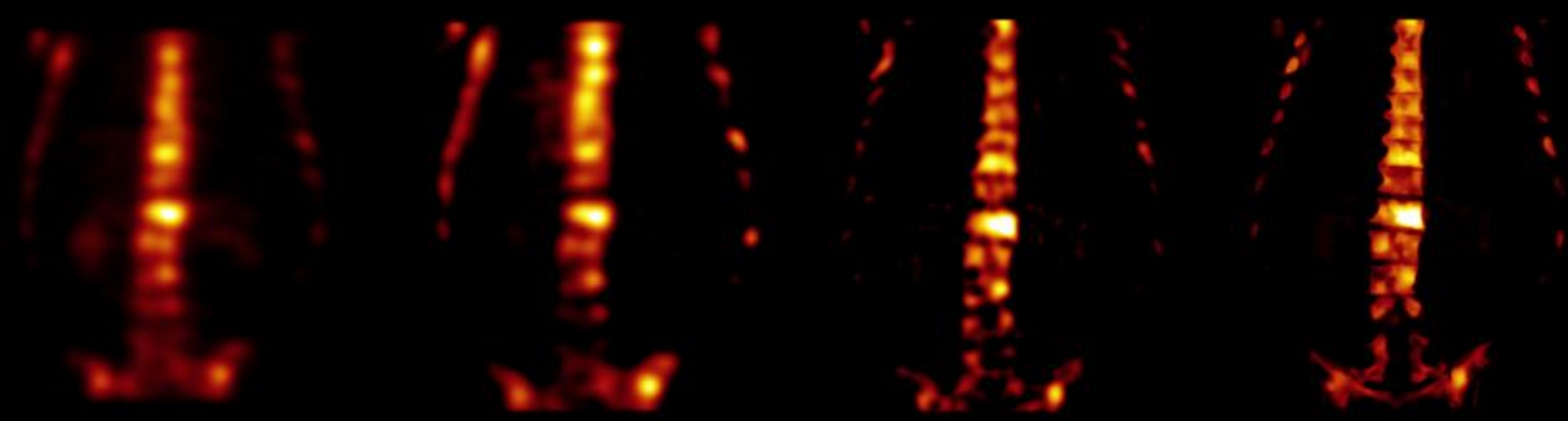
Phantom studies reconstructed with conventional 3D iterative reconstruction using CT attenuation correction in 256x256 matrix size and xSPECT reconstruction using 256x256 matrix size with CT attenuation correction. Data courtesy of Siemens Medical Solutions.

**See the Unseen**

Improved bone edge resolution for optimal visualization of vertebral metastases

SIEMENS

| Filtered-Back Projection | 3D Iterative with CT AC | xSPECT | xSPECT Bone* |
| --- | --- | --- | --- |



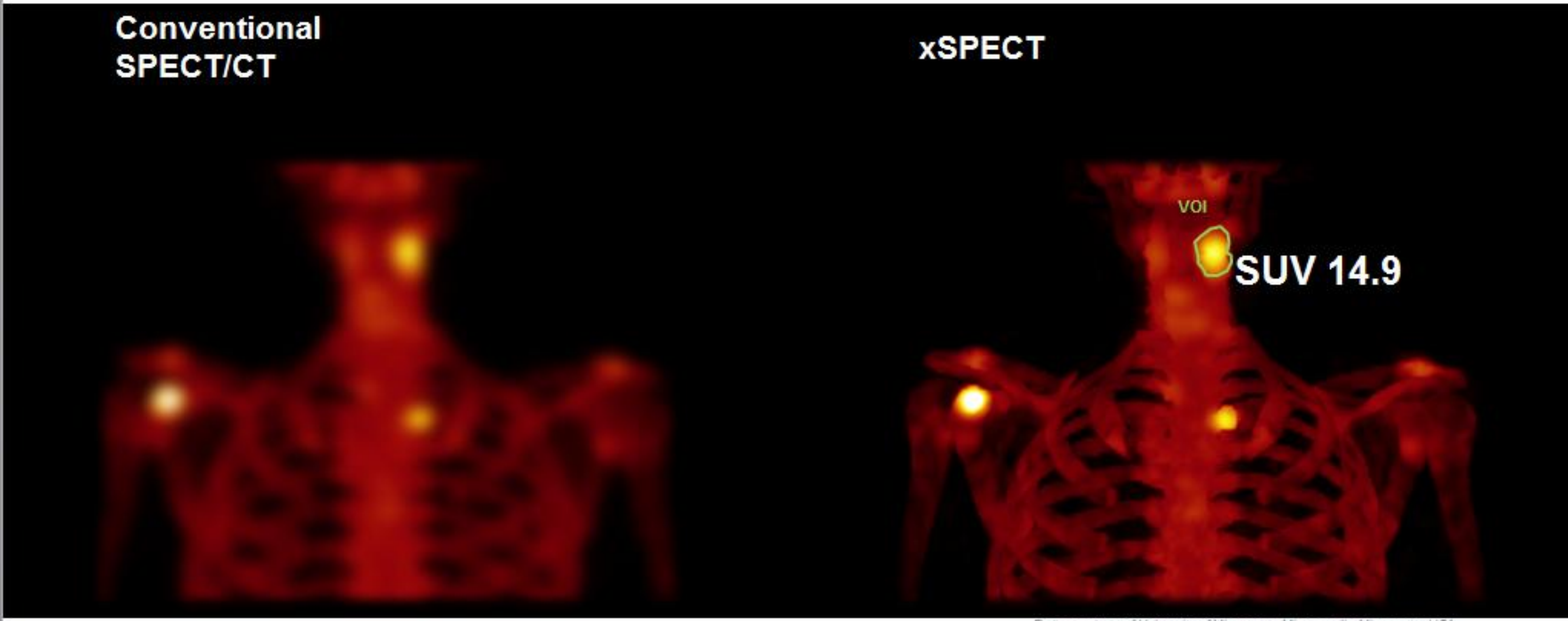
Data courtesy of University of Minnesota, Minneapolis, Minnesota, USA
Parameters: sex: female; weight: 85 kg (187 lbs); height 169 cm (5' 5");
injected dose: 929 MBq (25.10 mCi); 70mAs; 130 kV; slice thickness: 2.5 mm
* xSPECT Bone is not commercially available in all countries. Due to regulatory reasons its future availability cannot be guaranteed. Please contact your local

# Quantitative SPECT

SIEMENS

**Quantify the Difference**

xSPECT improves visual and quantitative assessment




Data courtesy of University of Minnesota, Minneapolis, Minnesota, USA



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# GE – Discovery NM/CT 670 Pro

## Discovery NM/CT 670 Pro

Discover what lies beyond the horizon.

Explore the deepest regions where disease arises

- Single breath-hold scans, 0.5 second rotation with high image quality
- 70cm chest abdomen pelvis in 10 seconds – IQE pitch booster covers more anatomy at the same image quality[2]
- Absolute quantitation of tracer uptake with Q.Metrix

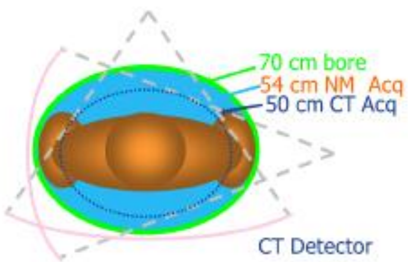

Overview | Design | SPECT | SPECT/CT | Quantitation | CT Applications

2

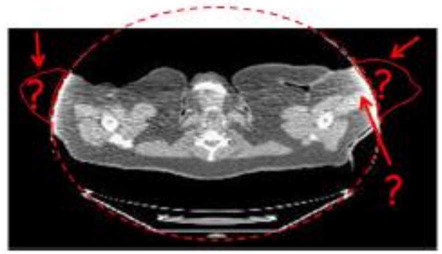
Diagnostic CT
Quantitative SPECT

# WideView CT for AC

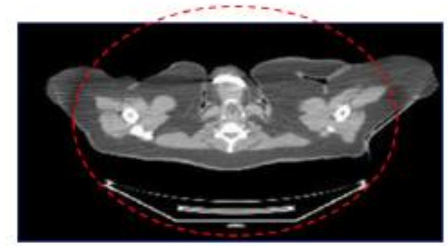
Removes CT clipping artifacts by completing truncated projections enabling attenuation correction throughout the entire SPECT FOV




Conventional 50 cm FOV CT Recon



Your challenge:
- Density inside the CT FOV is distorted close to the truncated edges
- Objects outs
are clipped



Our Solution[19]:
- Density inside the CT FOV is recovered
- Objects outside the CT FOV
stored

Overview | Design | SPECT | SPECT/CT | Quantitation | CT Applications

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Wide Bore System

# Advanced CT Technology

### Key Features

- 16 slices x 0.625 mm
- ASiR* reconstruction[14]
- 50 slices equivalent with IQE 1.75 pitch
- Powerful ergonomic operator console

### Key Benefits

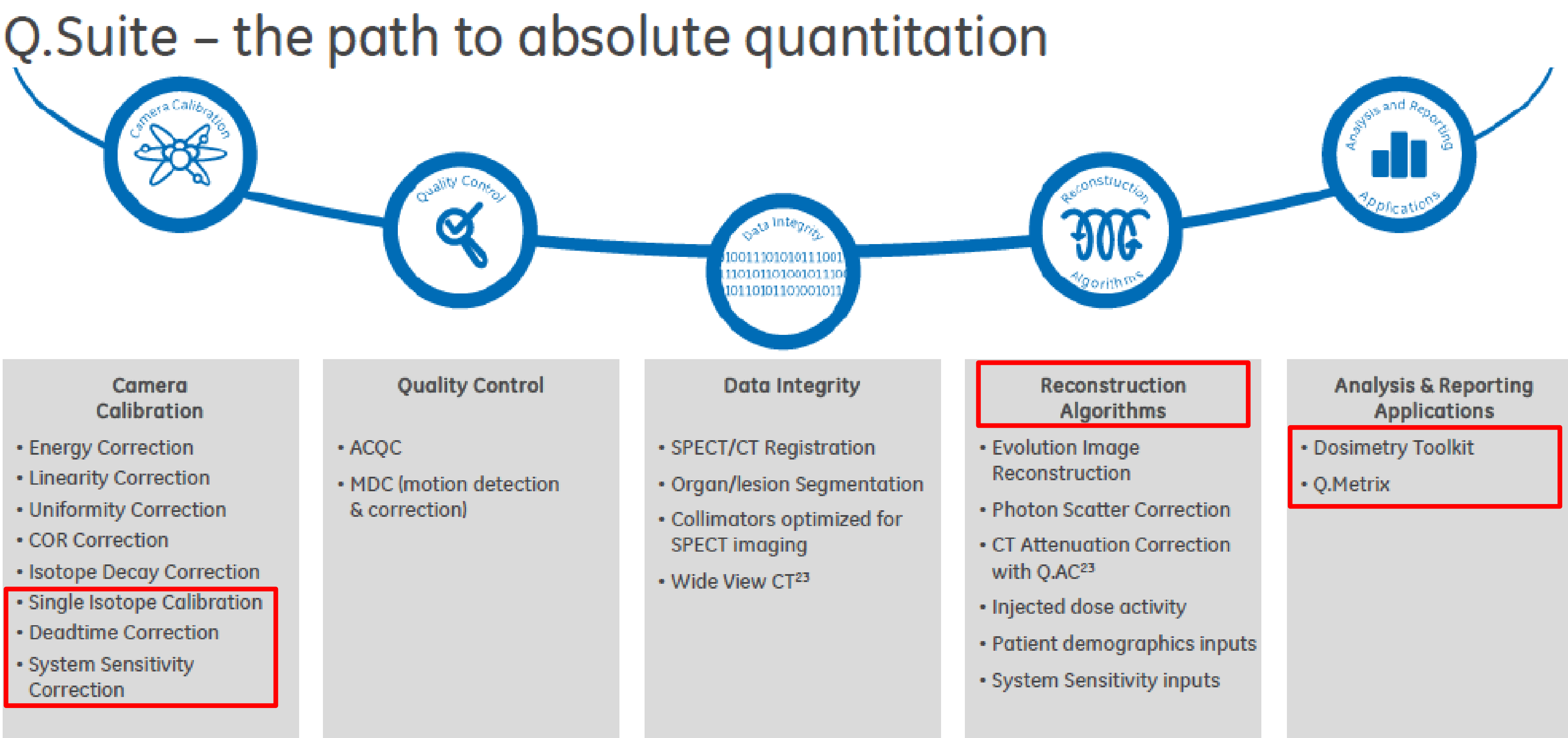
- Superb spatial resolution for the whole body
- Lower dose capabilities for patients of all ages
- Speed & coverage for time critical scans
- Efficient workflow

Scan range setting using **interactive ruler**




* Tra

# Quantitative SPECT

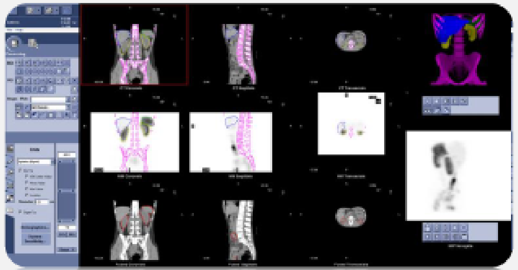



- Q.AC Low Dose CT Attenuation Correction Algorithm
  - Improved CT value accuracy at low mAs and/or kVp
- Advanced Application: ACQC, Volumetrix, Evolution,

# Q.Metrix : Absolute Quantitation

**Q.Metrix**

Q.Metrix employs SPECT and CT segmentation tools for quantifying radiopharmaceutical uptake in the form of MBq/ml. Using patient demographics information to calculate SPECT SUV with the same methods that are currently used to calculate SUV's for PET



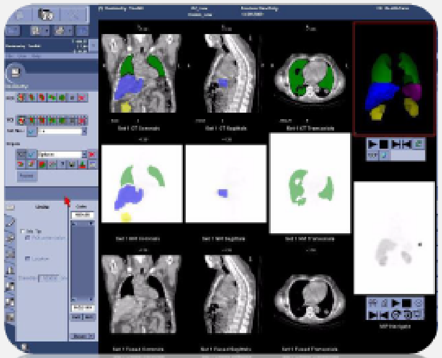
Quantification SPECT statistics calculated by Q.Metrix may be used for the following purposes:

- Calculate regional activity concentration (in MBq/ml)
- Define thresholds for different types of lesions using SPECT SUV values
- Study-to-study comparable statistics
- Possible follow-up and post treatment

# Dosimetry Toolkit: Absolute Quantitation

**Dosimetry Toolkit**

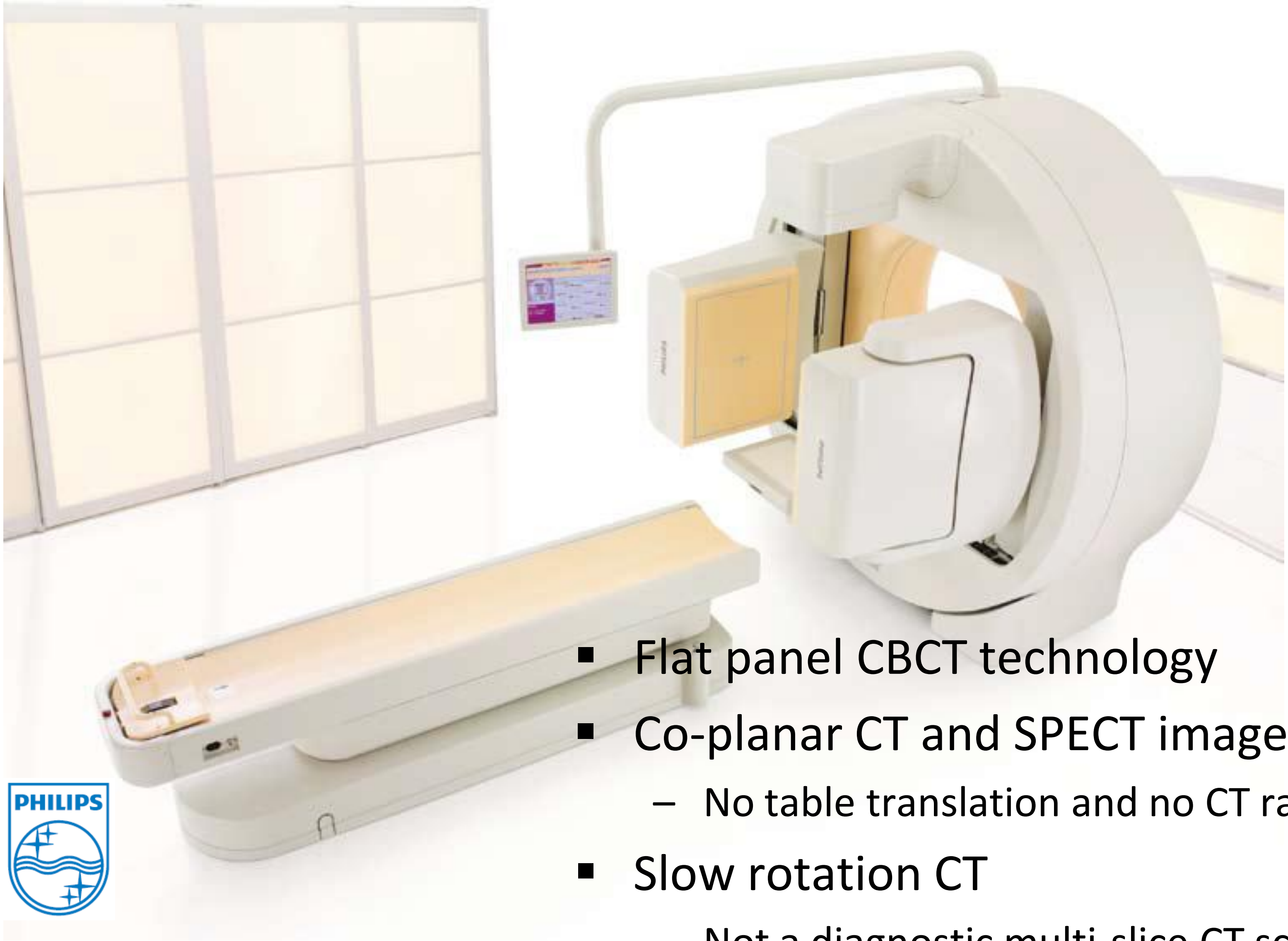
Used to define and report the patient organ volumes and activity, to quantify changes in radiopharmaceutical uptake over time and to calculate the residence time per organ.



These results can be based on the following types:

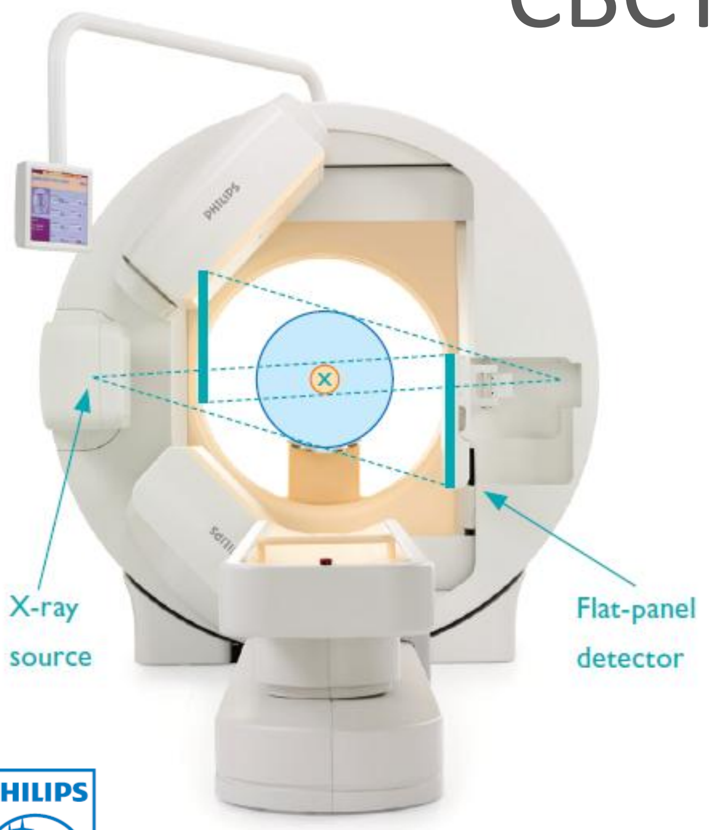
- Series of Multi Field of View SPECT/CT scans
- Series of whole body planar scans with a single SPECT/CT (Hybrid scenario)
- Series of planar WB (for which volume results can not be provided)

# Philips BrightView XCT



- Flat panel CBCT technology
- Co-planar CT and SPECT image acquisition
  - No table translation and no CT radiograph
- Slow rotation CT
  - Not a diagnostic multi-slice CT scanner

# CBCT Technology

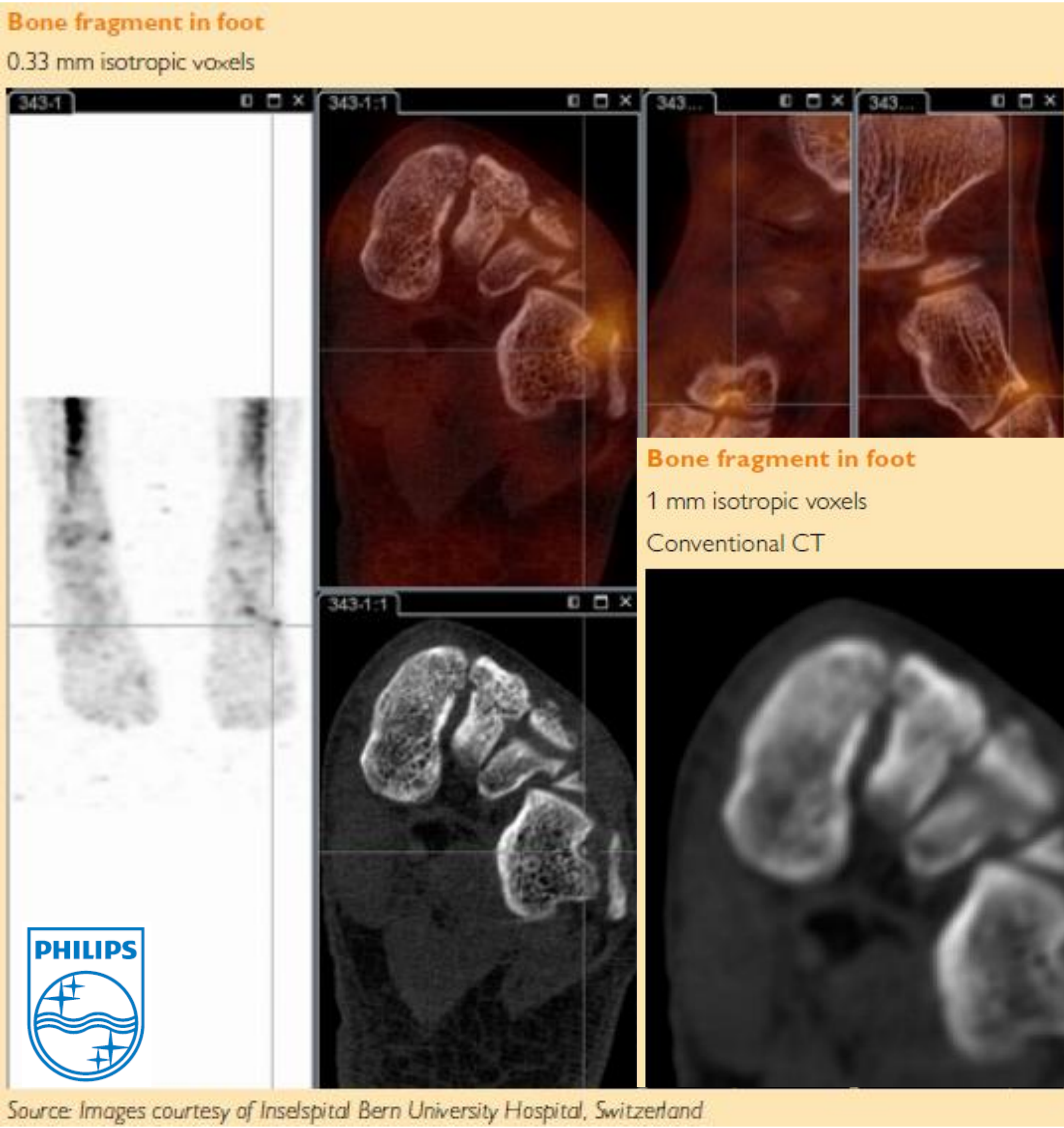


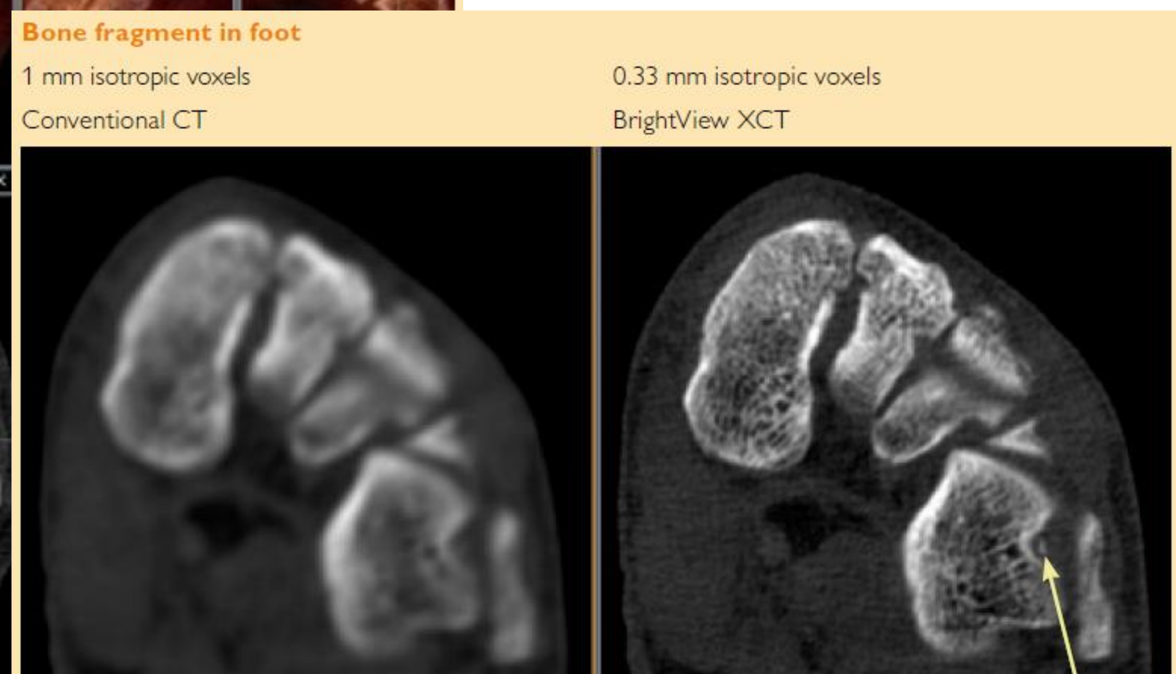

- FP is laterally offset from X-ray tube
- 1 X-ray projection covers slightly more than half of the CT FOV
- With 360° rotation, 47 cm diameter transverse FoV and a 14.4 cm axial length can be imaged
- 12, 24, or 60 second rotation times
- Co-planar CT and SPECT

| XCT physical assembly* | |
|---|---|
| Type of detector | Digital amorphous silicon, columnar CsI scintillator |
| Detector size | 30 cm x 40 cm |
| Detector pixel | 0.2 mm x 0.2 mm |
| Number of elements | 3,145,728 |
| Generator output | 10 kW, pulsed (2 msec. to continuous) |
| kVp | 120 rotating anode |
| mA | 5 – 80 |

| XCT performance* | |
|---|---|
| Axial field-of-view | 14 cm in a single 360° rotation |
| Maximum rotation speed | 12 seconds for 360° rotation |
| Maximum axial range | 172 cm |
| Transaxial field-of-view | 47 cm |
| Spatial resolution | > 15 lp/cm @ 10% MTF |

# High-Resolution CT images

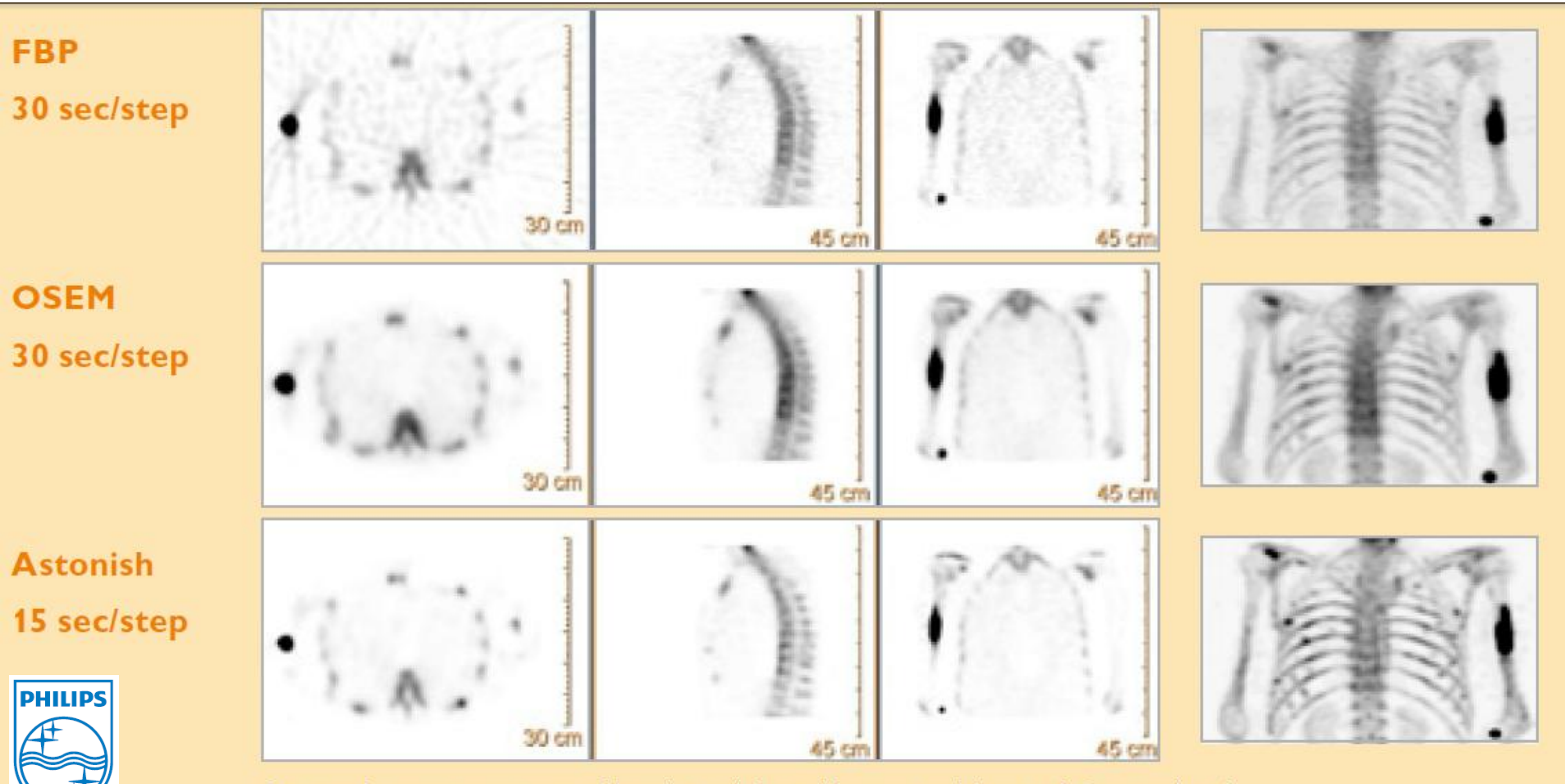



- **Isotropic voxel size**
- 1 mm for entire FOV
- 0.33 mm for subset-FOV
- SART Iterative Reconstruction

# Reconstruction: Astonish

- OSEM with 3D resolution recovery
- Patented noise-dampening technique – lower scan time




*Source: Images courtesy of Inselspital Bern University Hospital, Switzerland*

# STRATOS Dosimetry Solution

- Research software package for 3D voxel dose calculation using SPECT/CT and PET/CT data
- Allows for use a combination of 3D and planar scans

Registration
Segmentation
2D/3D data
User Calibrations
Dose Calculation
Evaluation Tools
TAC, DVH, VOI stats

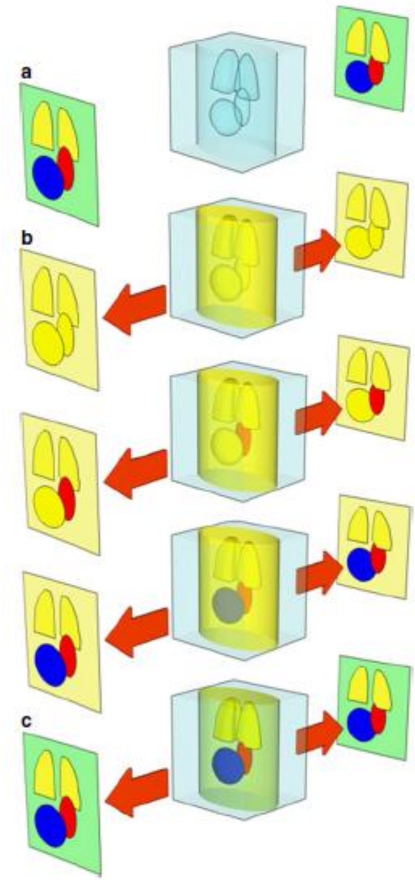
List of **supported tracers**

**Therapy** isotopes:

$^{18}$F, $^{90}$Y, $^{131}$I, $^{177}$Lu, $^{166}$Ho, $^{188}$Rh, $^{32}$P, $^{153}$Sm

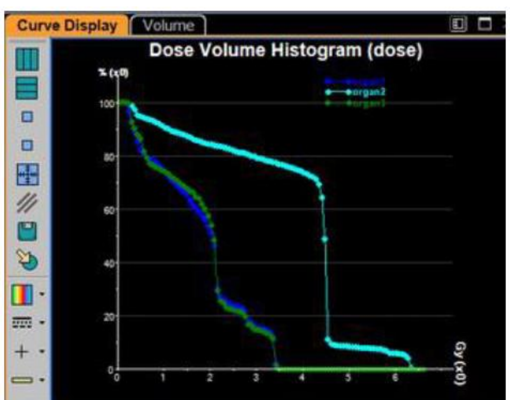
**Imaging** isotopes:

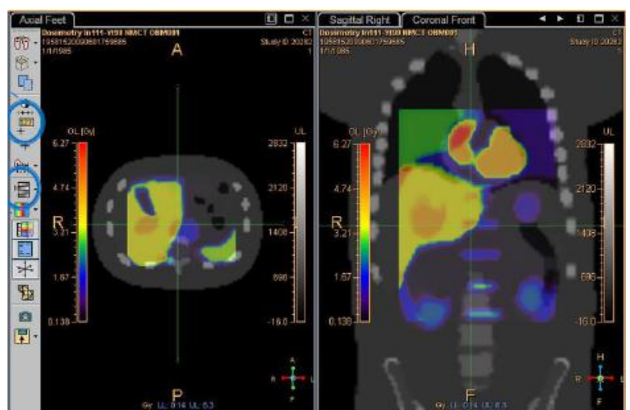
Therapy isotopes and $^{68}$Ga, $^{124}$I, $^{111}$In, $^{99}$Tc




PHILIPS

PHILIPS
Imalytics



# SAM Question 1

The most important function of the CT component of a hybrid SPECT/CT scanner is:

| | | |
|---|---|---|
| 0% | A. | Patient positioning in the SPECT scanner |
| 2% | B. | SPECT scatter correction |
| 98% | C. | Generation of μ-map for SPECT attenuation correction |
| 0% | D. | Enables faster SPECT scans |
| 0% | E. | Required for reconstruction of SPECT data |

# SAM Question 1: Answer

- The most important function of the CT component of a hybrid SPECT/CT scanner is:

  A. Patient positioning in the SPECT scanner
  B. SPECT scatter correction
  C. Generation of $\mu$-map for SPECT attenuation correction
  D. Enables faster SPECT scans
  E. Required for reconstruction of SPECT data

- **Answer: C**

- *Reference: SPECT/CT, Buck AK et al., J Nuclear Medicine 49(8), 1305-1319, 2008*
- *Reference: Investigation of the use of x-ray CT images for attenuation correction in SPECT, LaCroix KJ et al., IEEE Trans Nuclear Science 41(6), 2793-2799, 1994*

# SAM Question 2

Iterative reconstruction techniques (e.g., OS-EM) are routinely used for reconstruction of SPECT emission data from hybrid SPECT/CT systems because:

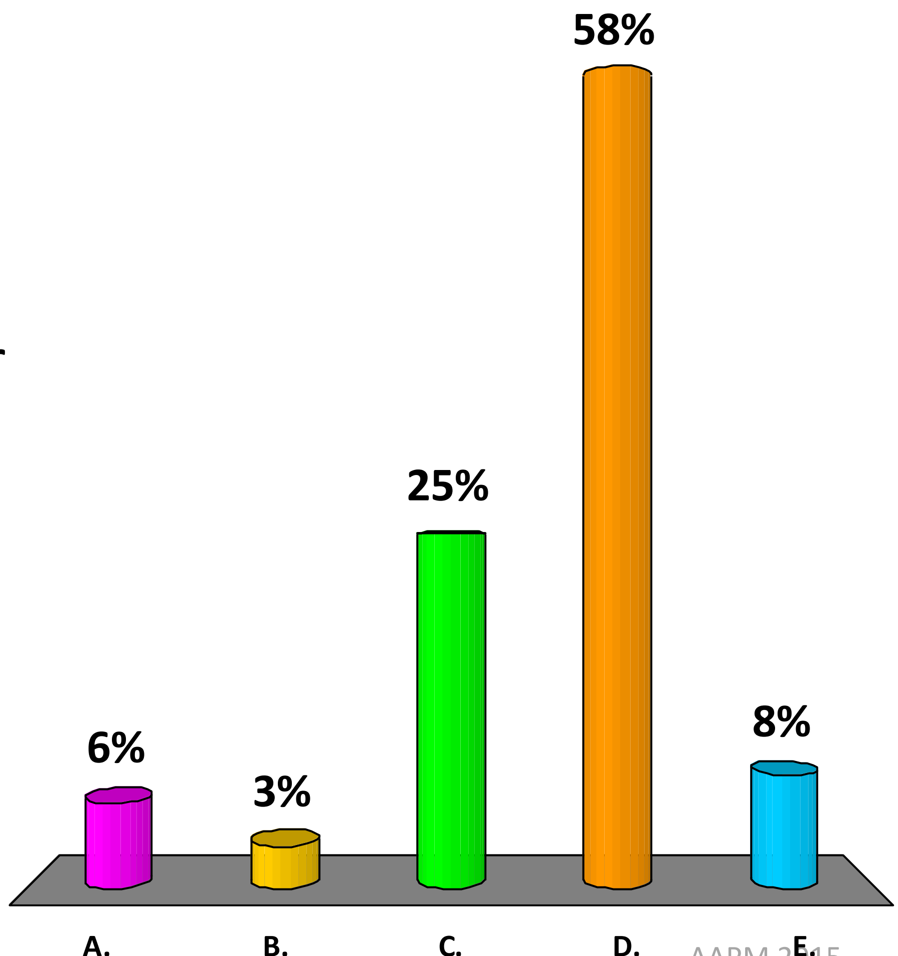
A. They are not affected by scatter

B. They are not affected attenuation correction

C. They require shorter computer processing time than FBP

D. They can accurately model the physics of gamma camera photon detection

E. They require CT images for image registration



# SAM Question 2: Answer

- Iterative reconstruction techniques (e.g., OS-EM) are routinely used for reconstruction of SPECT emission data from hybrid SPECT/CT systems because:
  A. They are not affected by scatter
  B. They are not affected attenuation correction
  C. They require shorter computer processing time than FBP
  D. They can accurately model the physics of gamma camera photon detection
  E. They require CT images for image registration

- **Answer: D**


- *Reference: Maximum likelihood reconstruction for emission tomography, Shepp LA and Vardi Y, IEEE Trans Medical Imaging 1, 113-122, 1982*
- *Reference: Quantitative analysis in nuclear medicine imaging, Zaidi H (editor), Springer New York, 2006*

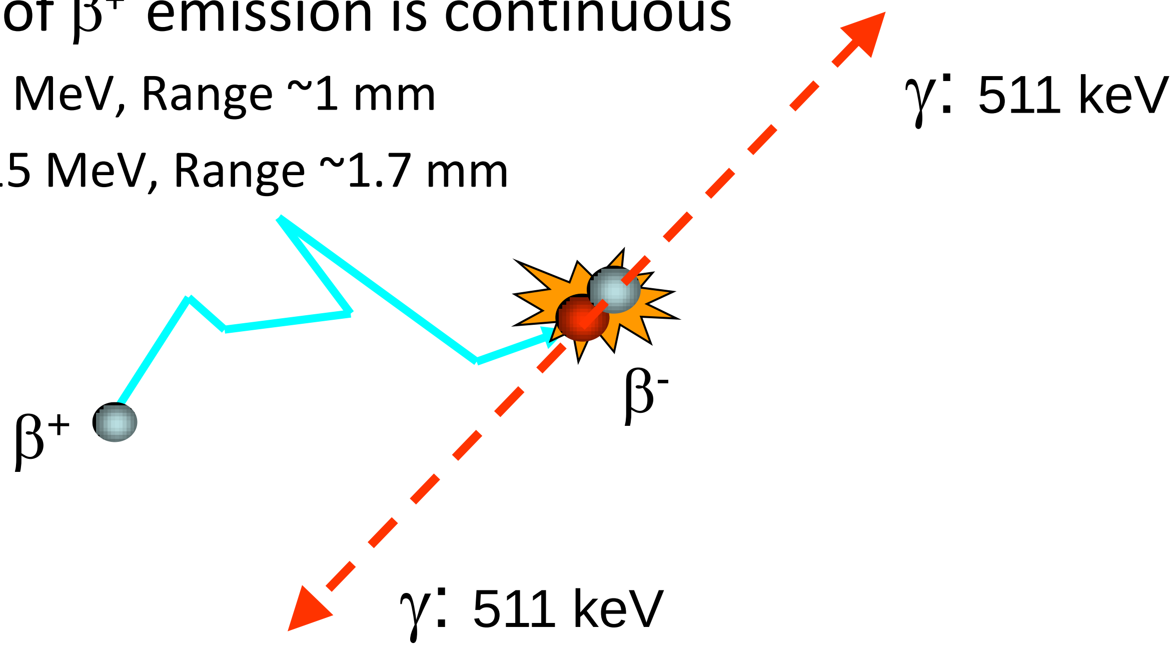
# PET/CT

# Annihilation Photons

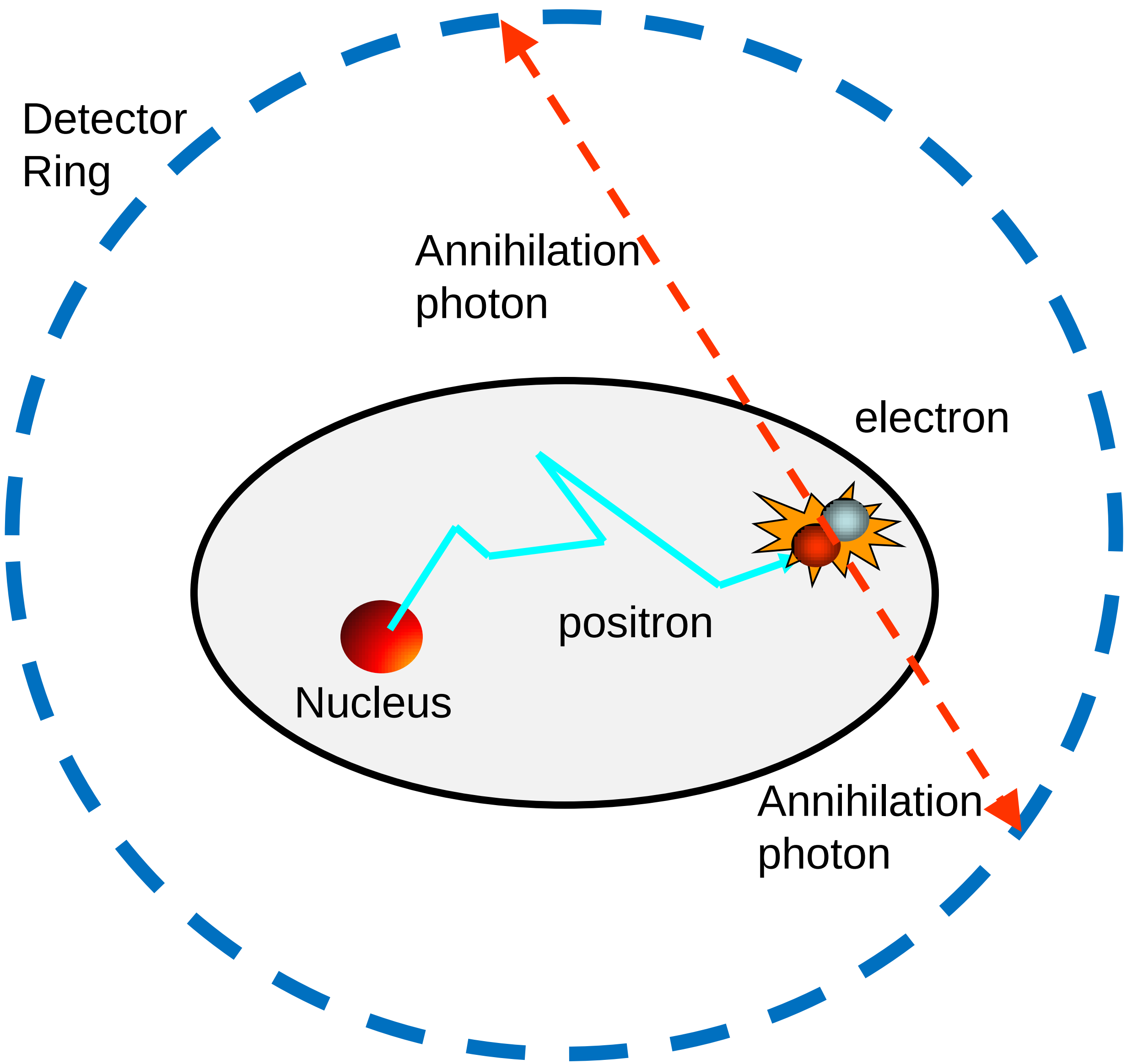
- Nuclei with low a neutron-to-proton ratio converts a proton to a neutron via emission of positron ($\beta^+$)

$$p = n + \beta^+ + \upsilon \; ; \; {}^{A}X_{Z} = {}^{A}Y_{Z-1} + \beta^+ + \upsilon$$

- Cyclotron (generator) for production of $\beta^+$ emitters
- $\beta^+$ annihilation → two simultaneous 511 keV photons
  - Emitted (nearly) 180 degrees apart
- Energy spectrum of $\beta^+$ emission is continuous
  - F18: Emax = 0.64 MeV, Range ~1 mm
  - Ru82: Emax = 3.15 MeV, Range ~1.7 mm

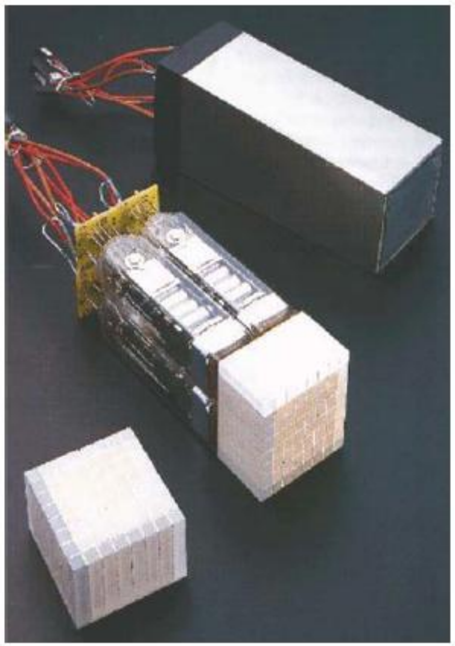


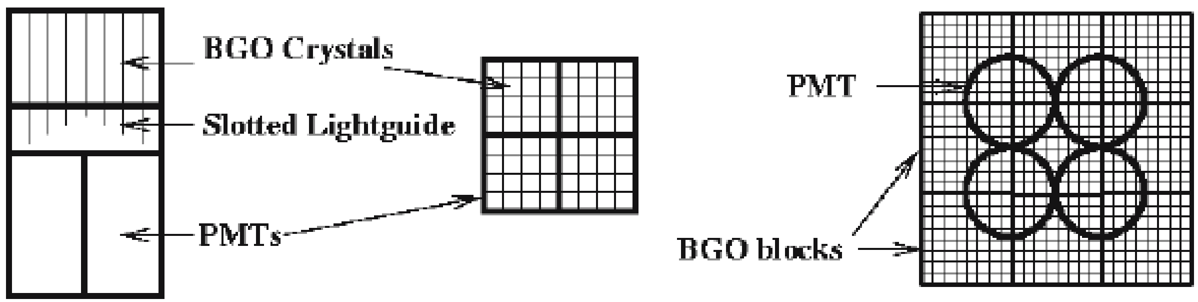
# Schematic of a PET scanner

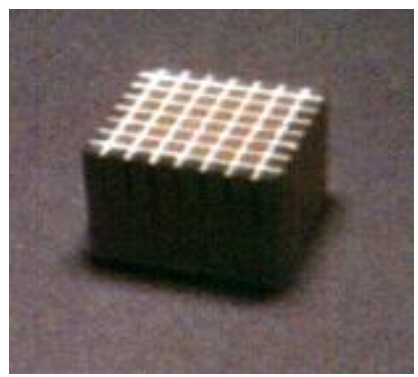


# PET detectors

| Scintillator | Relative light output [NaI(Tl)=100] | Decay time (ns) | Thickness for 90% efficiency at 511 keV (cm) |
|---|---|---|---|
| BGO | 15 | 300 | 2.4 |
| GSO | 25 | 60 | 3.3 |
| LSO, LYSO | 80 | 40 | 2.7 |

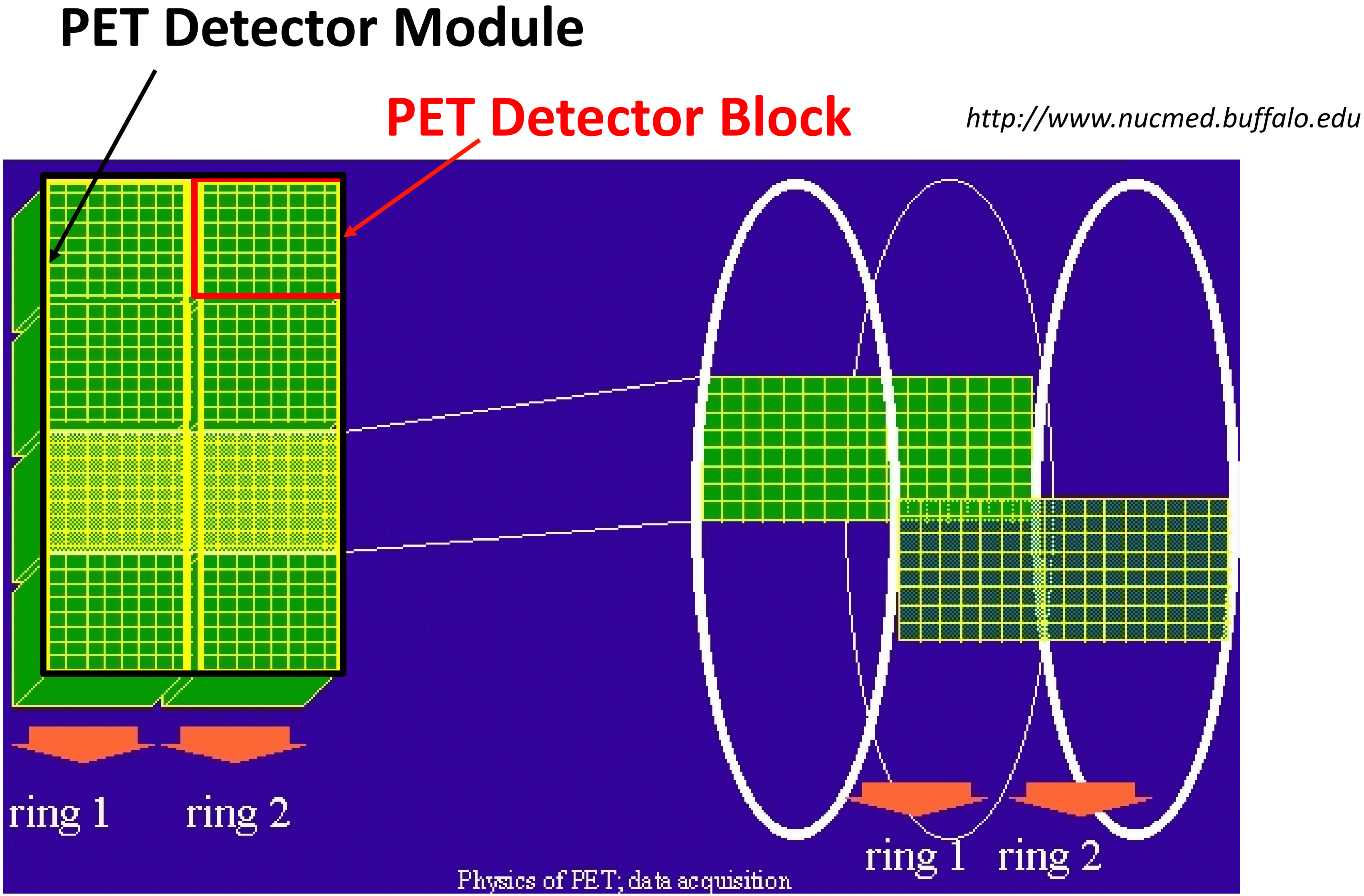




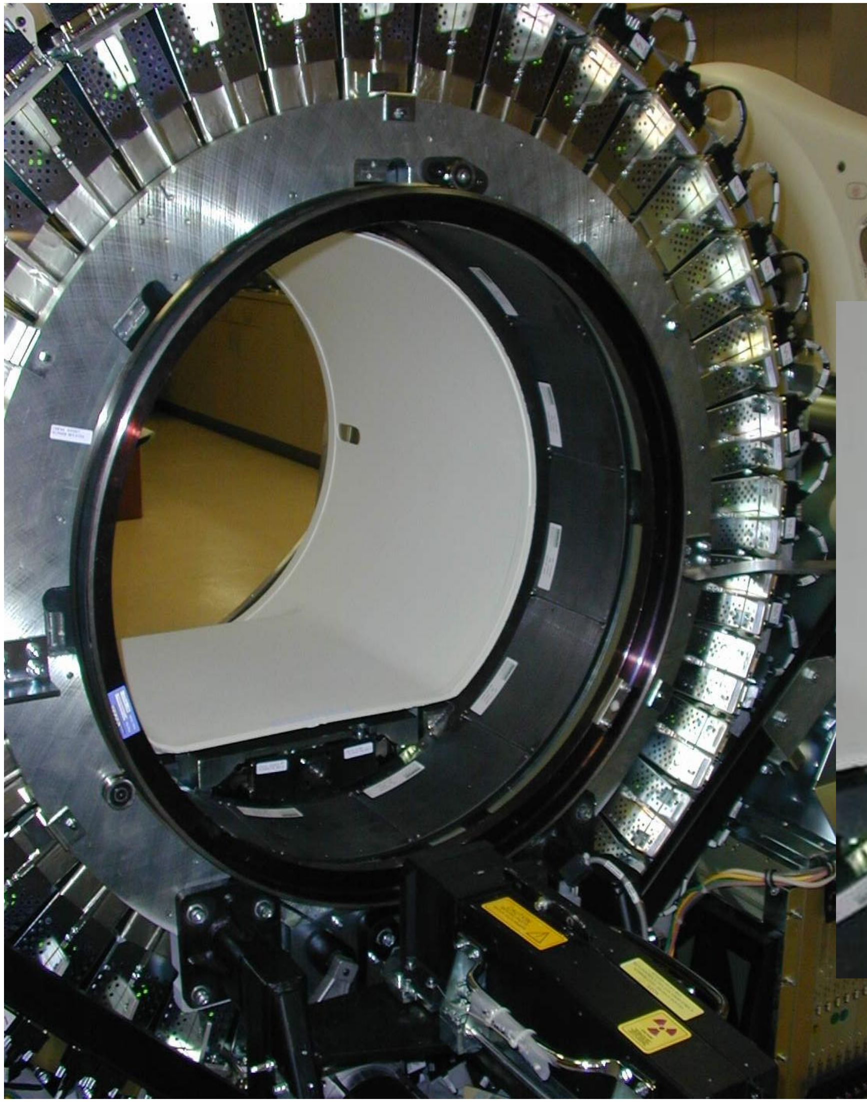



**PET Detector Block**

# PET Detector Module and Rings

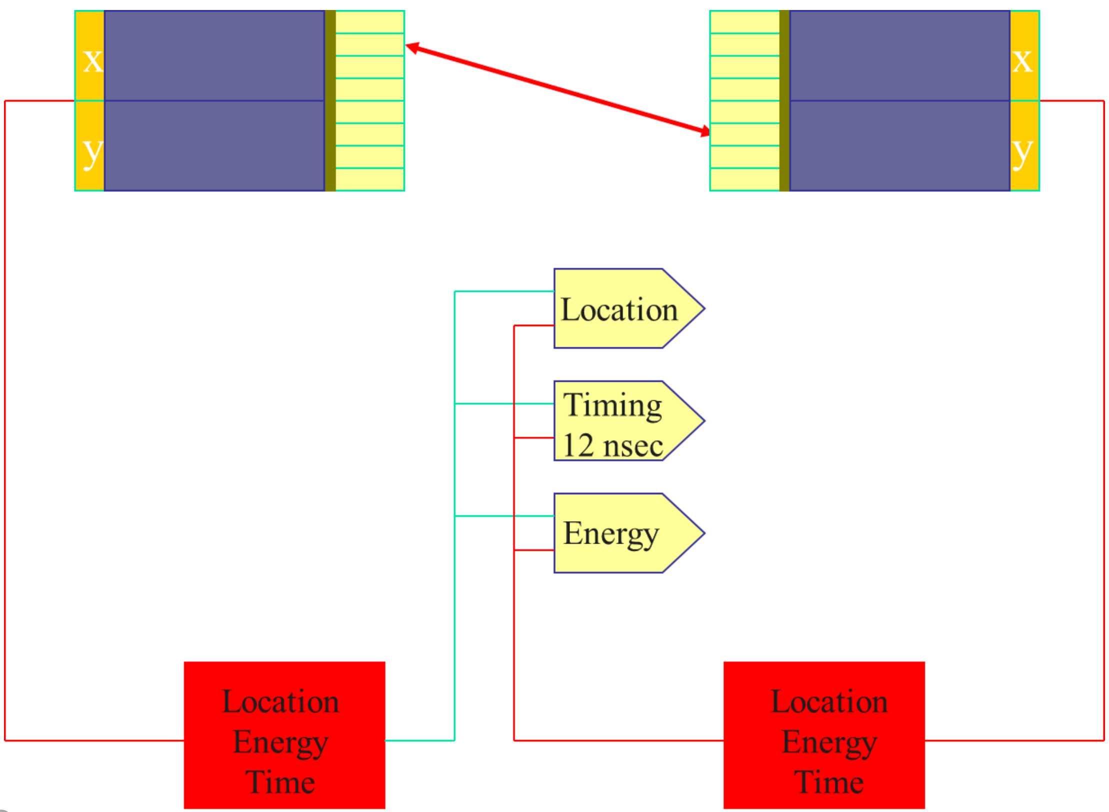


# PET Scanner – Covers Off

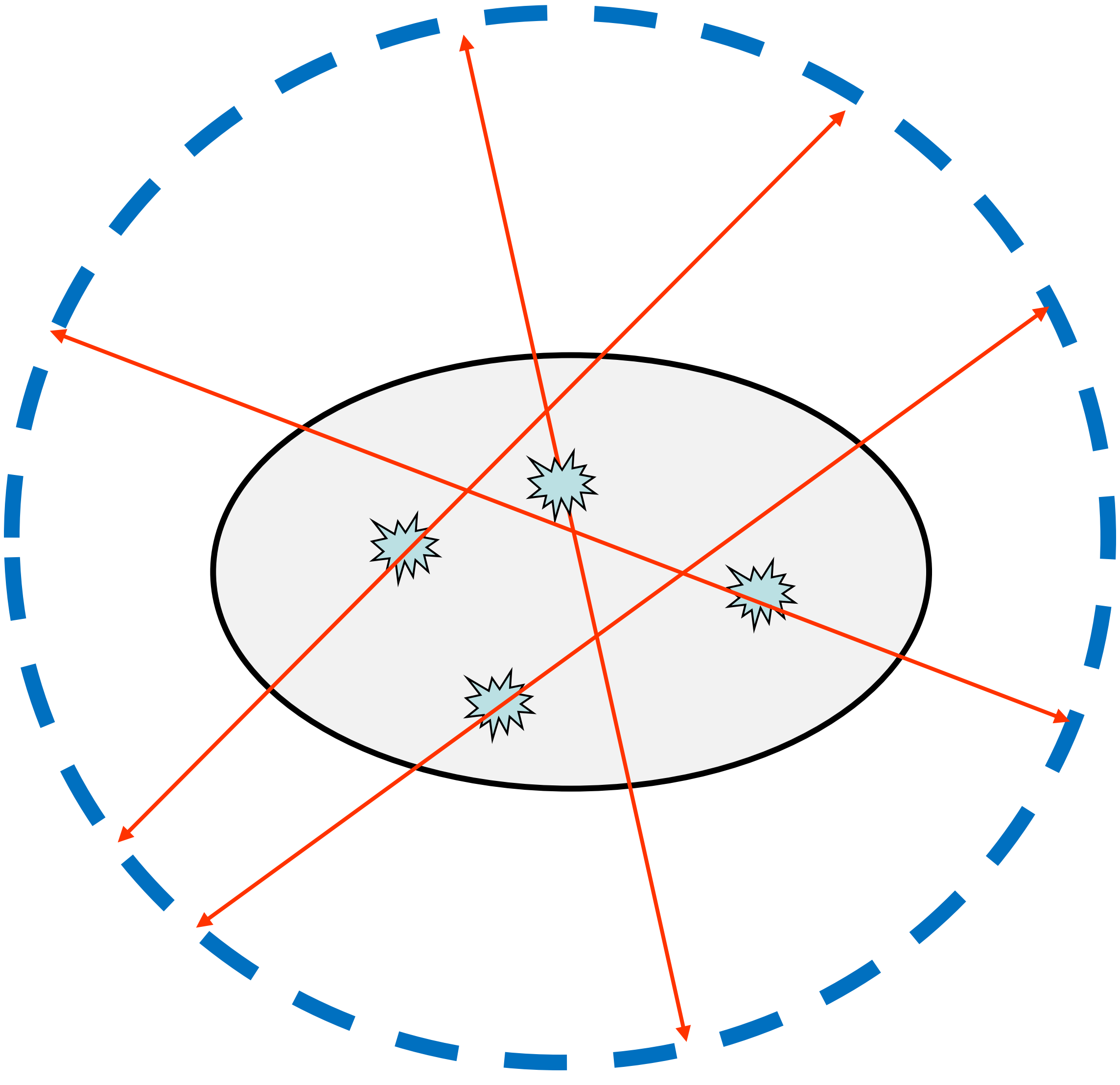


# Record the Line-of-Response

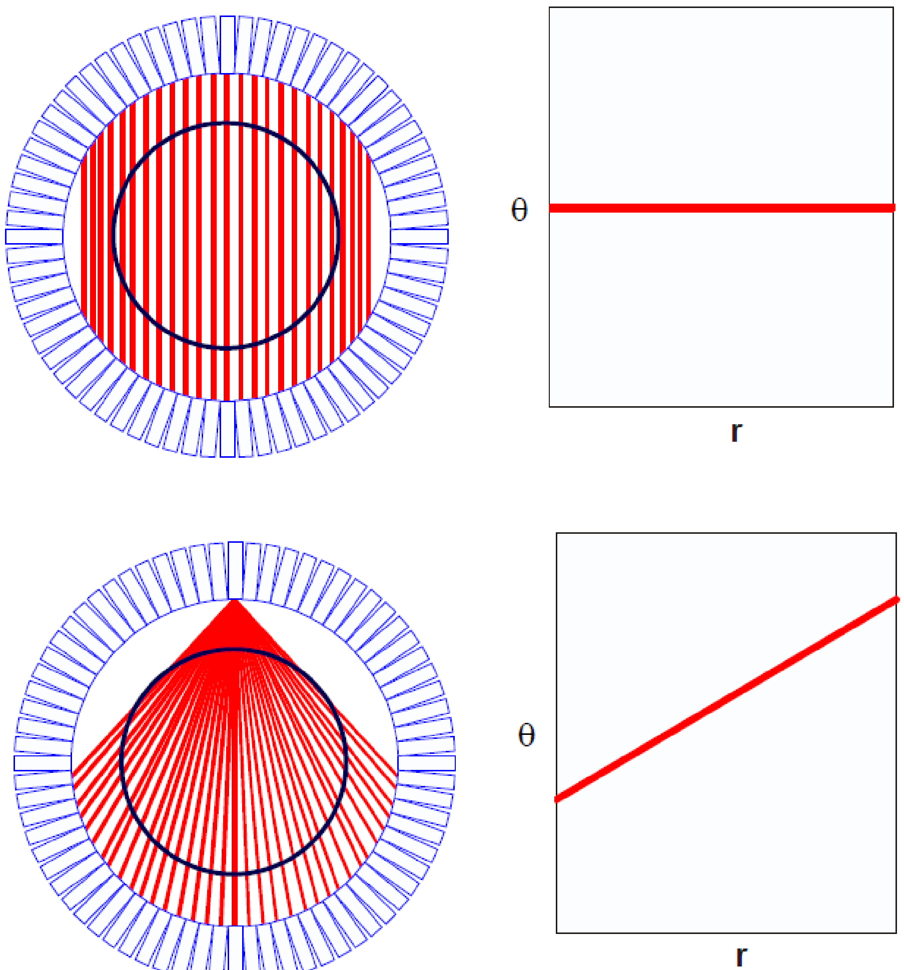
- Fundamental prerequisite to PET imaging
  - Photon (Singles) detection and processing
  - Coincidence assessment of singles events
  - Data storage and processing



# PET Detector Ring



# LOR to Sinograms



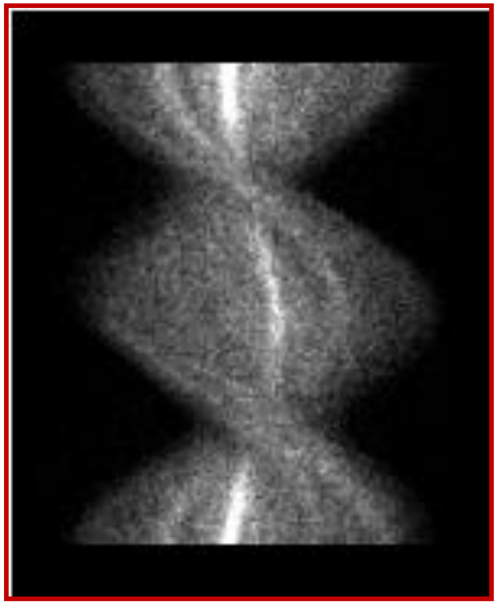
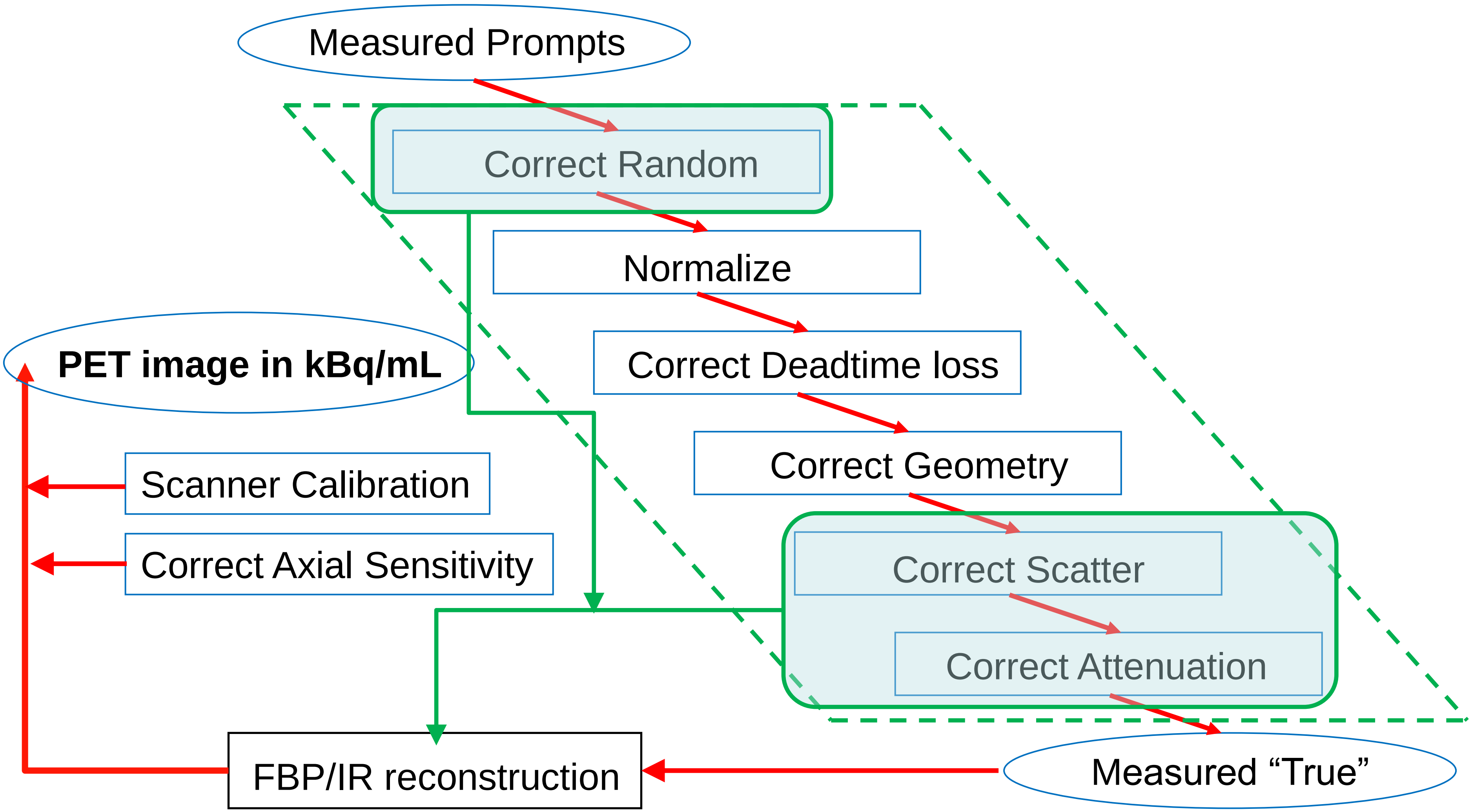
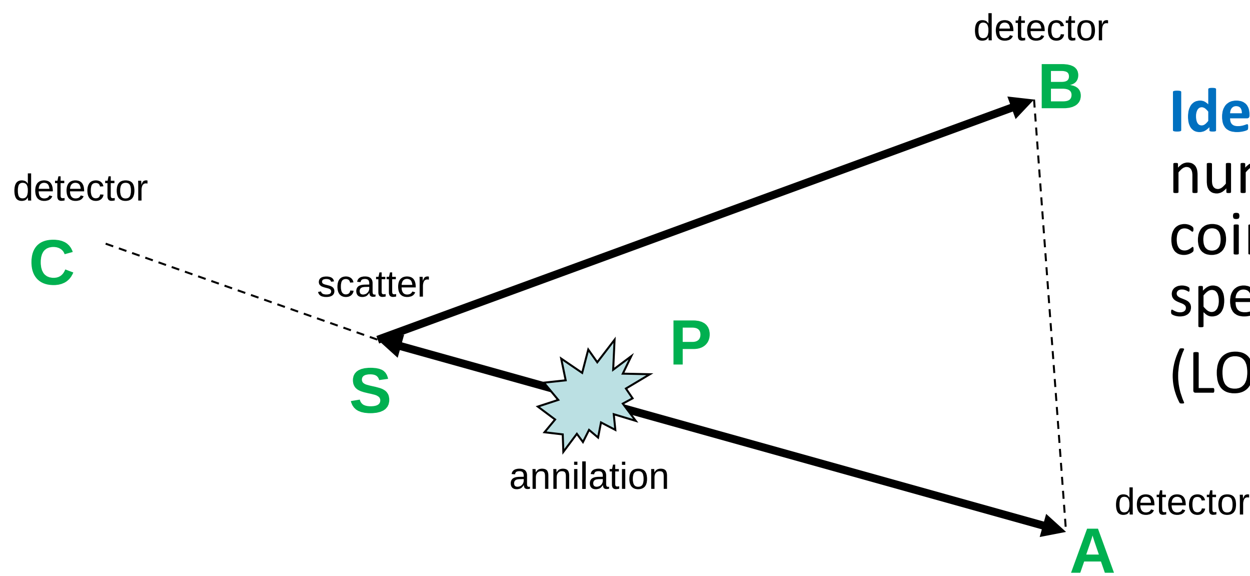



*Image Courtesy: Magnus Dahlbom*

# PET data corrections



# Model-based Scatter Estimation




**Idea**: To estimate the number of scattered coincidence along a specific LOR

(LOR AB in figure)

Ollinger, Phys. Med. Biol. 41 (1996) 153-176

Assume an annihilation at point P,
- Compute probability the photons originate along AC
- Compute the probability that the one of the photon is detected at A
- Compute the probability of second photon scattering at location S
- Compute the fraction of events scattered toward B (Klein-Nishina formula)
- The probability that the scattered photon is detected at B

**Input: PET emission image, CT transmission image, LOR AB**
**Output: Scatter along LOR AB**

# PET Scanner Calibration

- Perform PET scan with low known activity
  - Low scatter and deadtime conditions
  - Uniform cylinder – simple attenuation correction
- Convert PET true count rate (cps) into activity concentration (Bq/mL)
- PET Standard Uptake Values $\left[\frac{Bq/mL}{Bq/mg}\right]$

$$\text{SUV} = \frac{\text{decay-corrected dose/ml of tumor}}{\text{injected dose/patient weight in grams}}$$

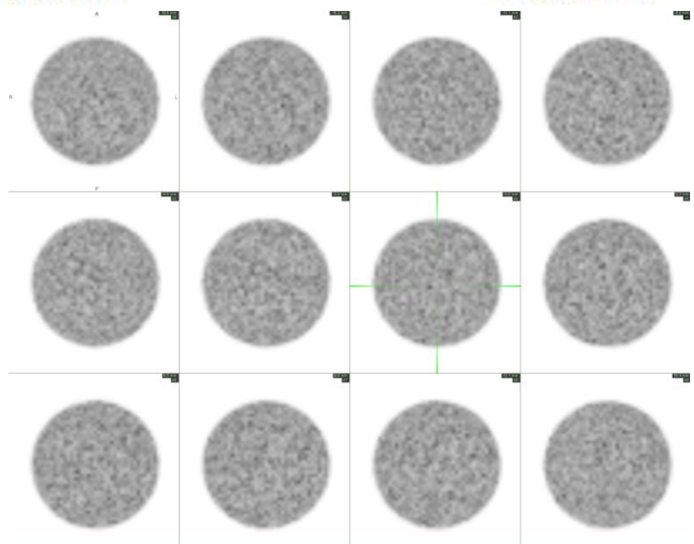
$$\text{SUV}_{\text{lean}} = \frac{\text{decay} - \text{corrected dose/ml of tumor}}{\text{injected dose/patient lean body mass in grams}}$$

# PET Calibration Phantoms

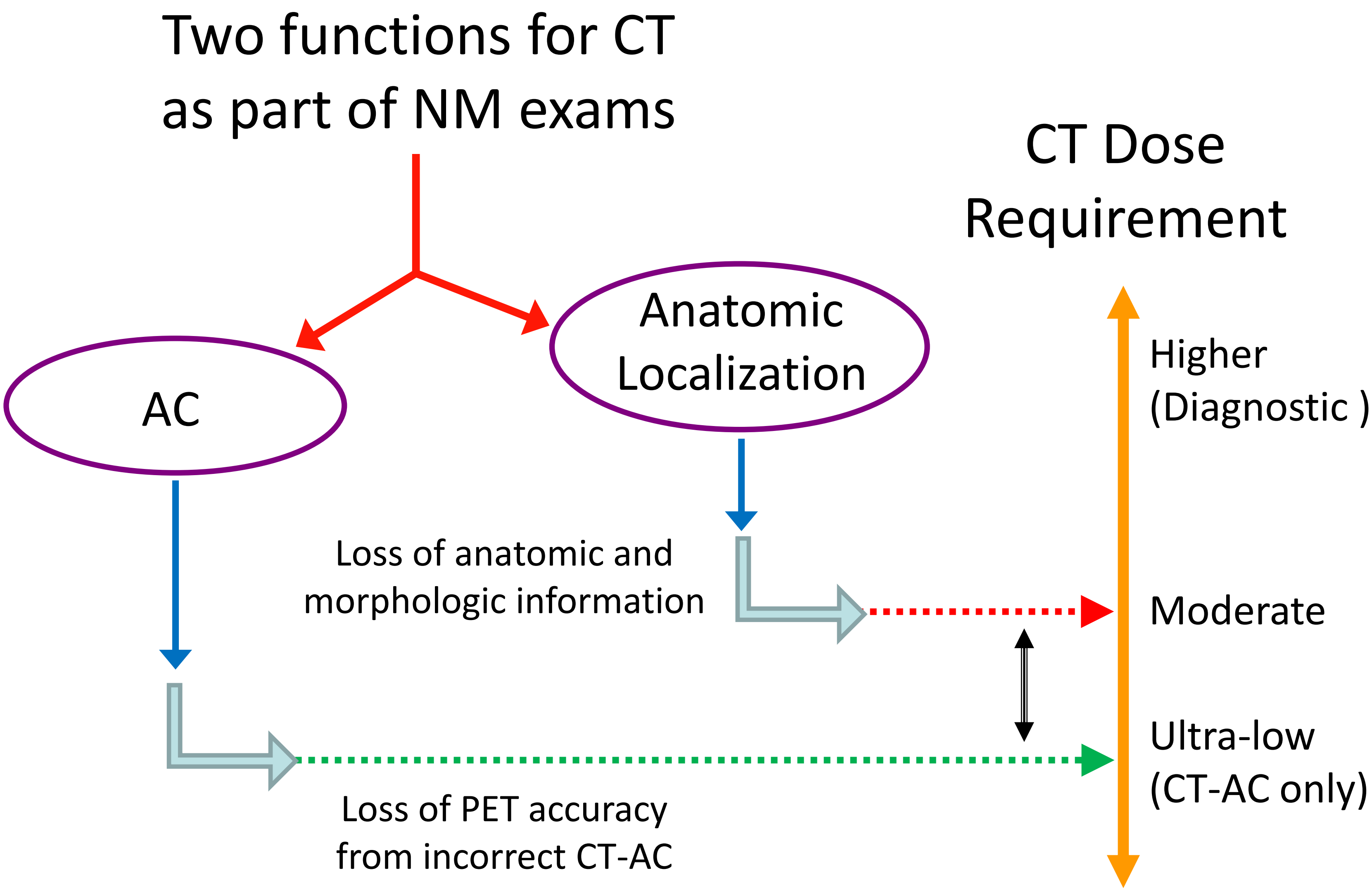
Water Phantom

Solid $^{68}$Ge Phantom

NIST traceable F-18 STD
"S" vial geometry



# Role of CT in PET/CT and SPECT/CT

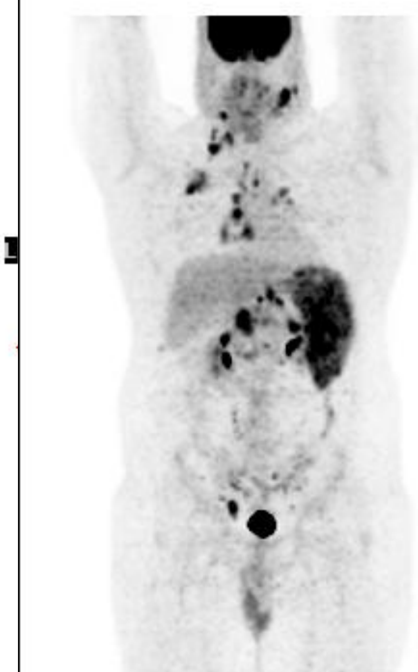


# PET/CT w/ and w/o AC

CT


PET w/o CT-AC

PET with CT-AC



Fused PET/CT

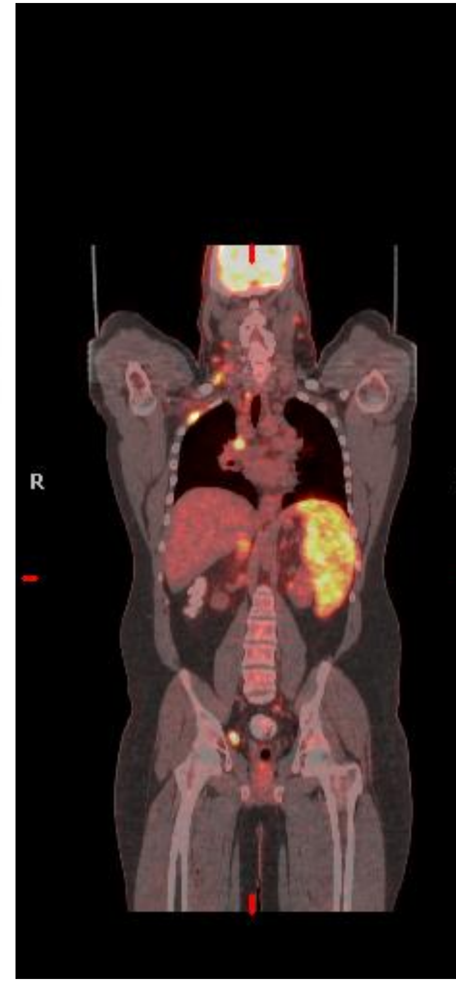


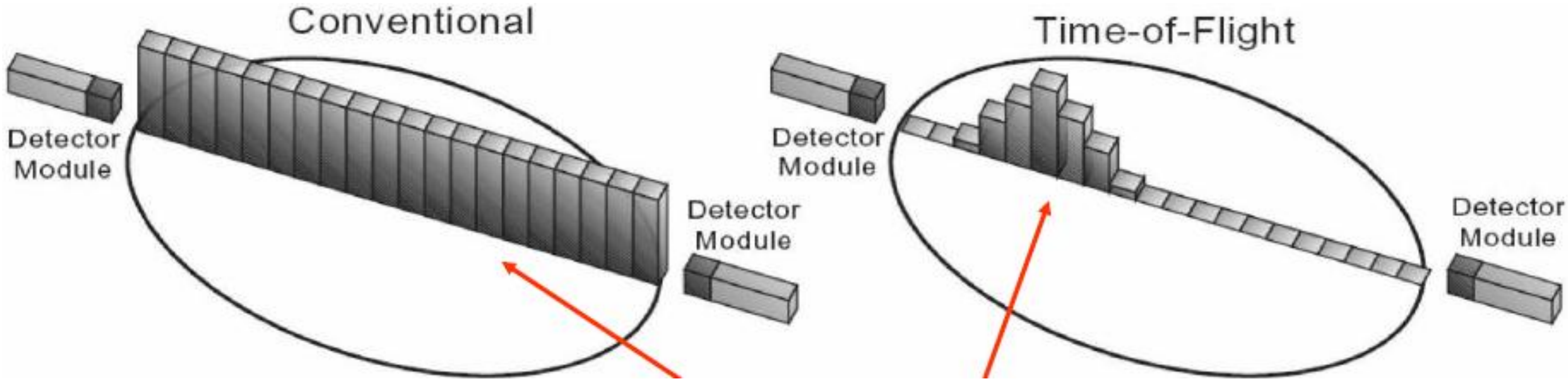

*Image Courtesy: Osama Mawlawi*

# Recent advances in PET/CT

- Recent advances
  - TOF PET
  - PSF modeling
  - Extended axial FOV
  - Gating for motion correction

- More recent advances
  - Continuous bed motion (Siemens FlowMotion)
  - Regularized reconstruction (GE Q.Clear)
  - Digital detectors (Phillips Vereos)

# Time-of-Flight PET




Probability along LOR

| $\Delta$t (ps) | $\Delta$x (cm) |
|---|---|
| 600 | 9 |
| 100 | 1.5 |
| 33 | 0.5 |

$$\Delta x = \frac{\Delta t}{2} c$$

$$SNR_{TOF} \cong \sqrt{\frac{D_{obj}}{\Delta x}} SNR_{non-TOF}$$

# TOF PET has higher Image Contrast

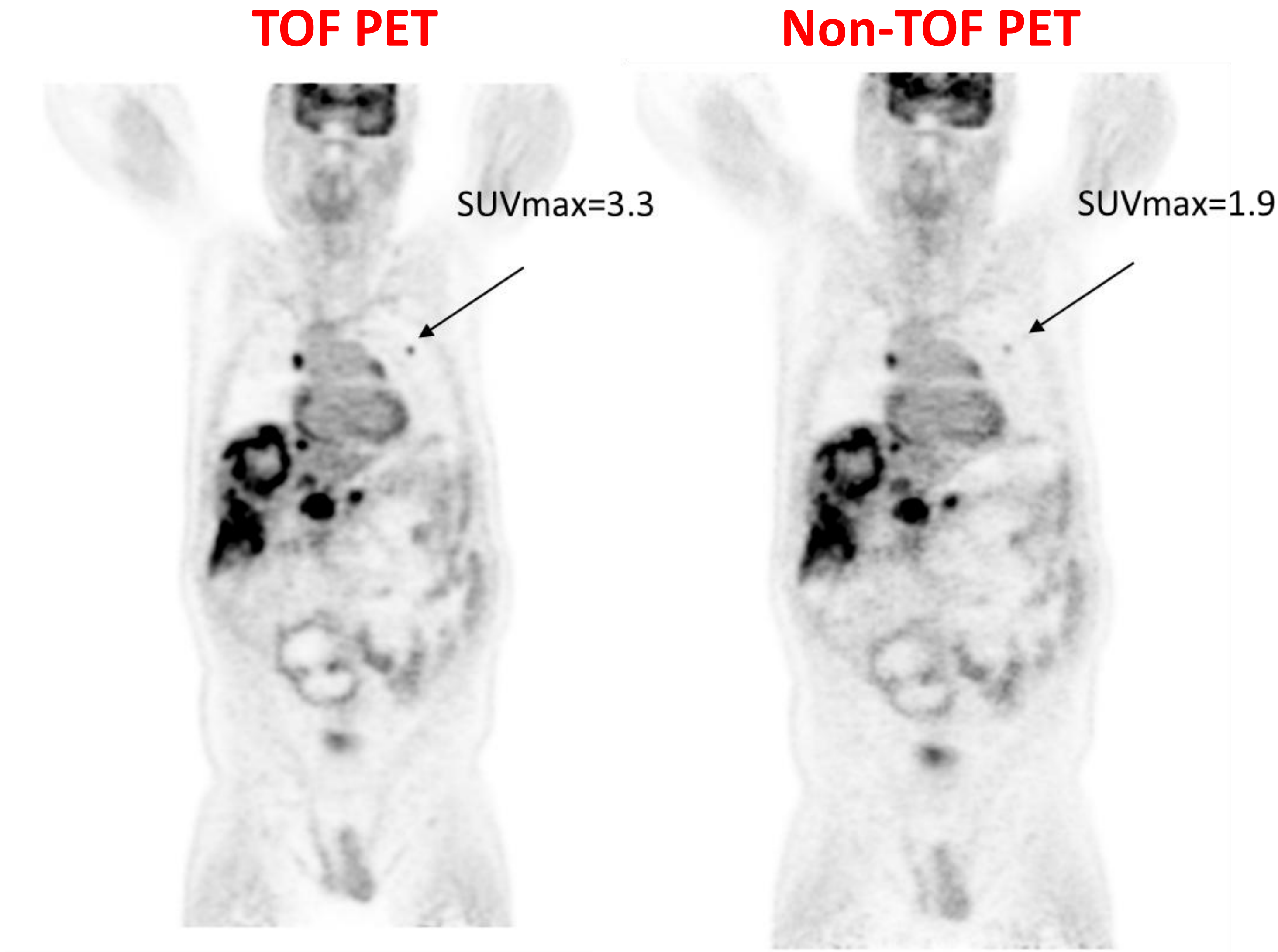
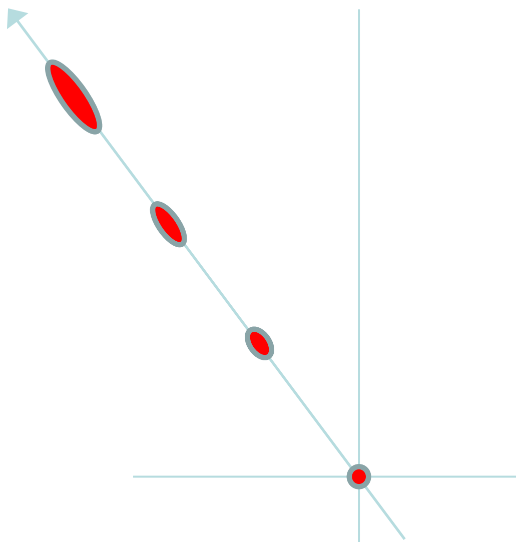
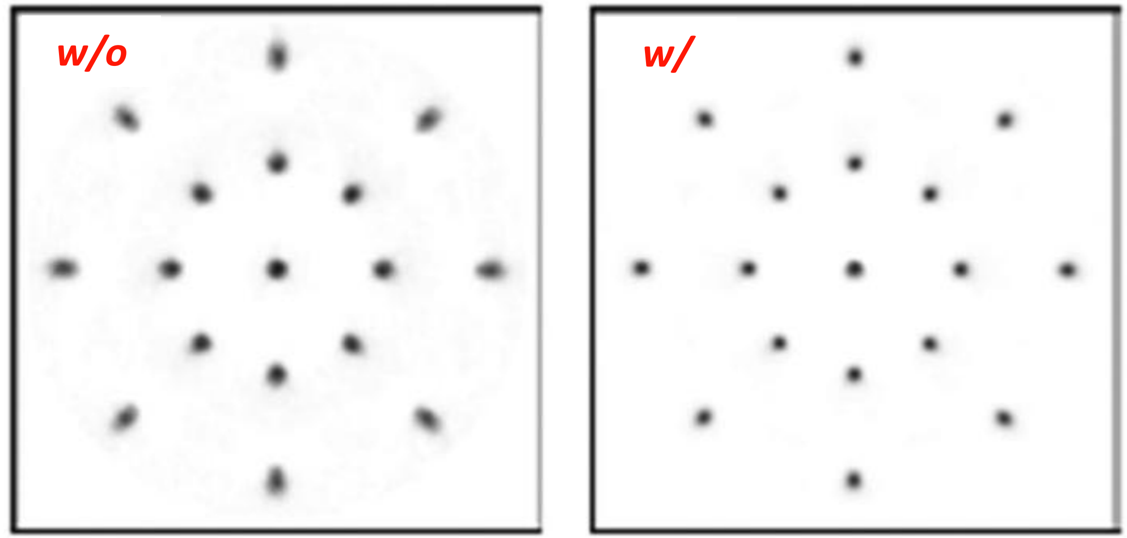



*Image Courtesy: Osama Mawlawi*

# PSF Resolution Modeling

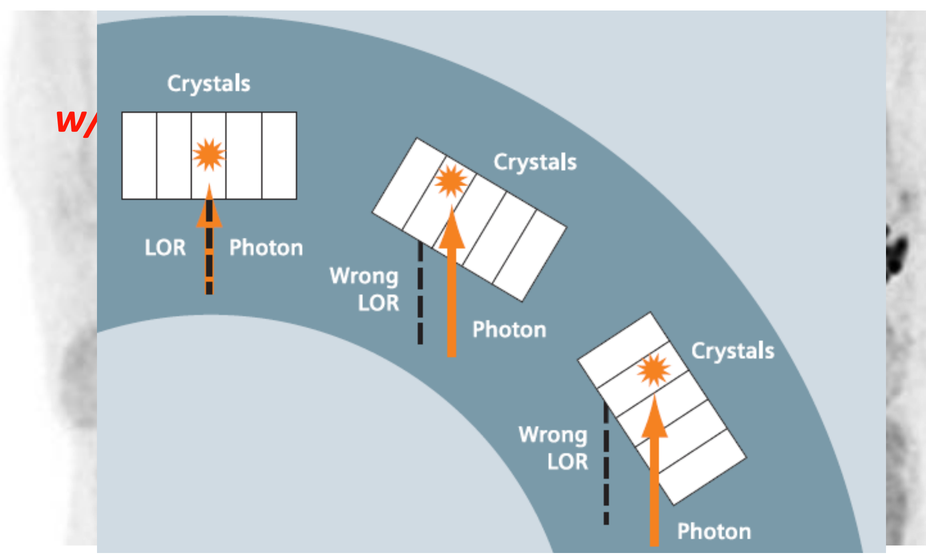


*Lee et al., PMB 49, 2004*




*Pecking et al., Clin. Exp. Metastasis 29, 2012*

- Goal is to improve image quality, contrast, and quantitative accuracy
- SharpIR (GE)
- TrueX (Siemens)
- Philips ✓



# PET Image Quality w/ PSF modelling

**WITH**

**WITHOUT**

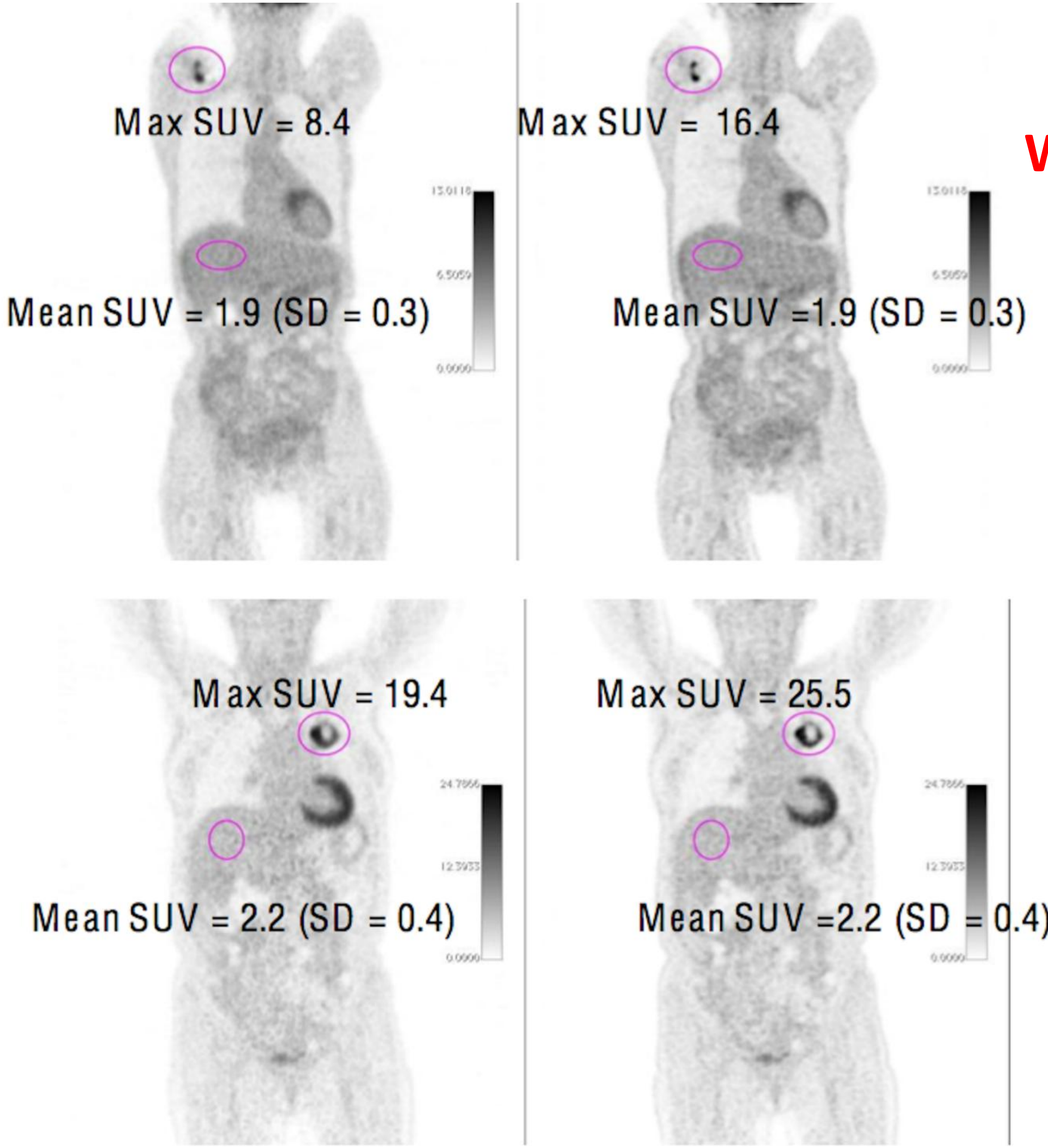



*Image Courtesy:*
*Osama Mawlawi*

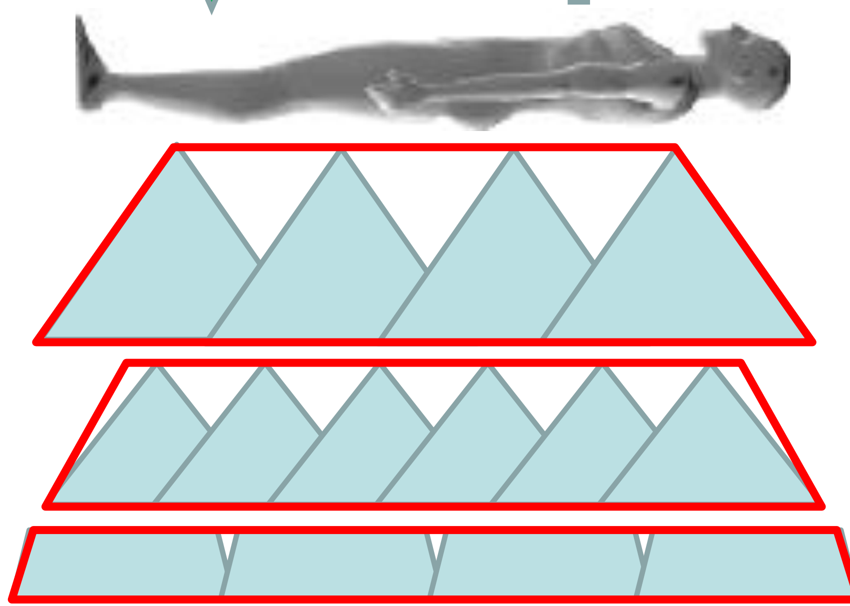
# 2D versus 3D PET

- 2D: Septa present between detector planes in axial direction
  - Reduces scatter; Uniform AX sensitivity; Small (~1 cm) bed overlap
- 3D:  No collimation present except at end of ring
  - Triangular AX sensitivity profile (~50% detector overlap)
  - Sensitivity 3D > 2D → lower activity needed
- 3D: Extended Axial FOV
  - Fewer bed positions for same axial coverage
  - Increased sensitivity → time/bed ↓ or counts/time ↑

**3D ext. Ax FOV**: Even Higher Sensitivity + Lower No. of Beds

**3D PET**: Higher Sensitivity + Greater No. of Beds

**2D PET**: Lower Sensitivity + Fewer No. of Beds

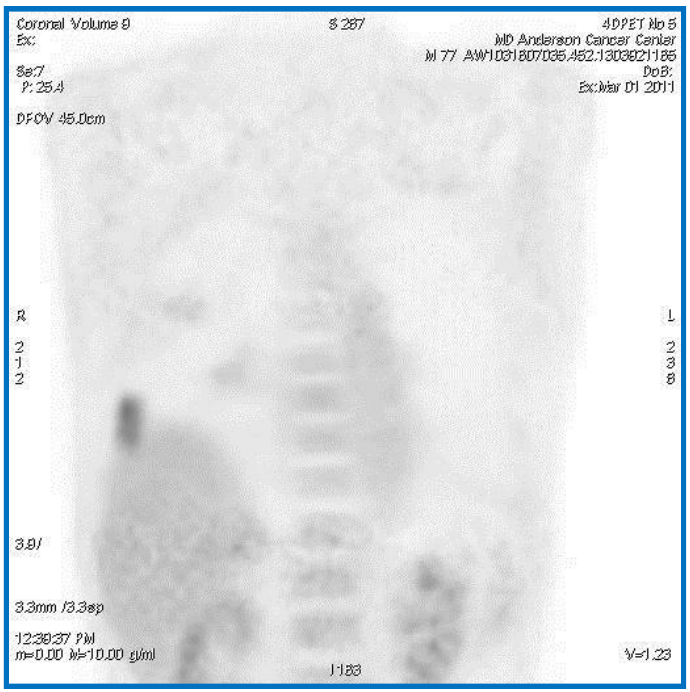


# Extended Axial FOV

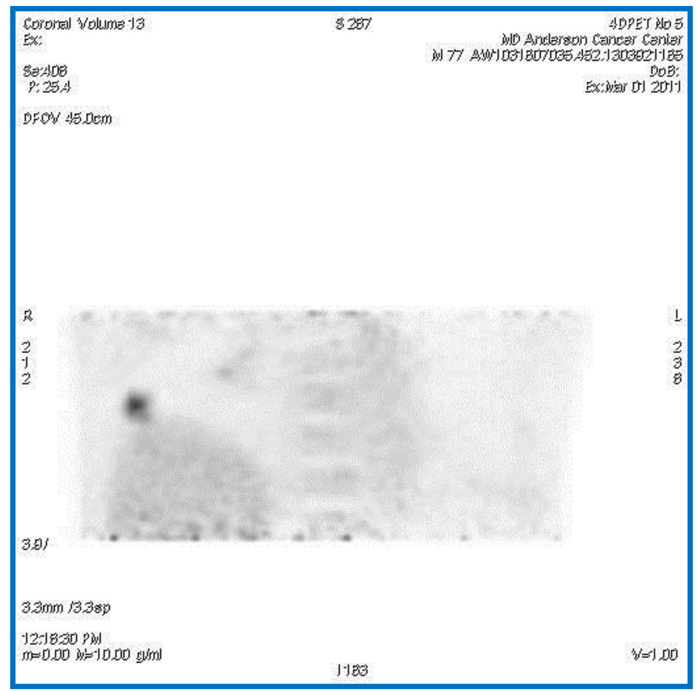
- Typical configuration
  - aFOV of 15-16 cm with Sensitivity of 5-7 cps/kBq
- GE: Discovery IQ (BGO, non-TOF)
  - aFOV options (cm): 15.5 to 26
  - Sensitivity (cps/kBq) = 7.5 to 22
- Siemens: Biograph mCT (LYSO, TOF)
  - aFOV options (cm): 16.2 to 21.6
  - Sensitivity (cps/kBq) = 5.5 to 10

# Gating and List Mode

- Motion smears PET signal and reduced intensity
  - PET is motion averaged therefore use (motion) average CT
- Trigger to sort PET data into bins to correct for organ motion – cardiac or respiratory gating



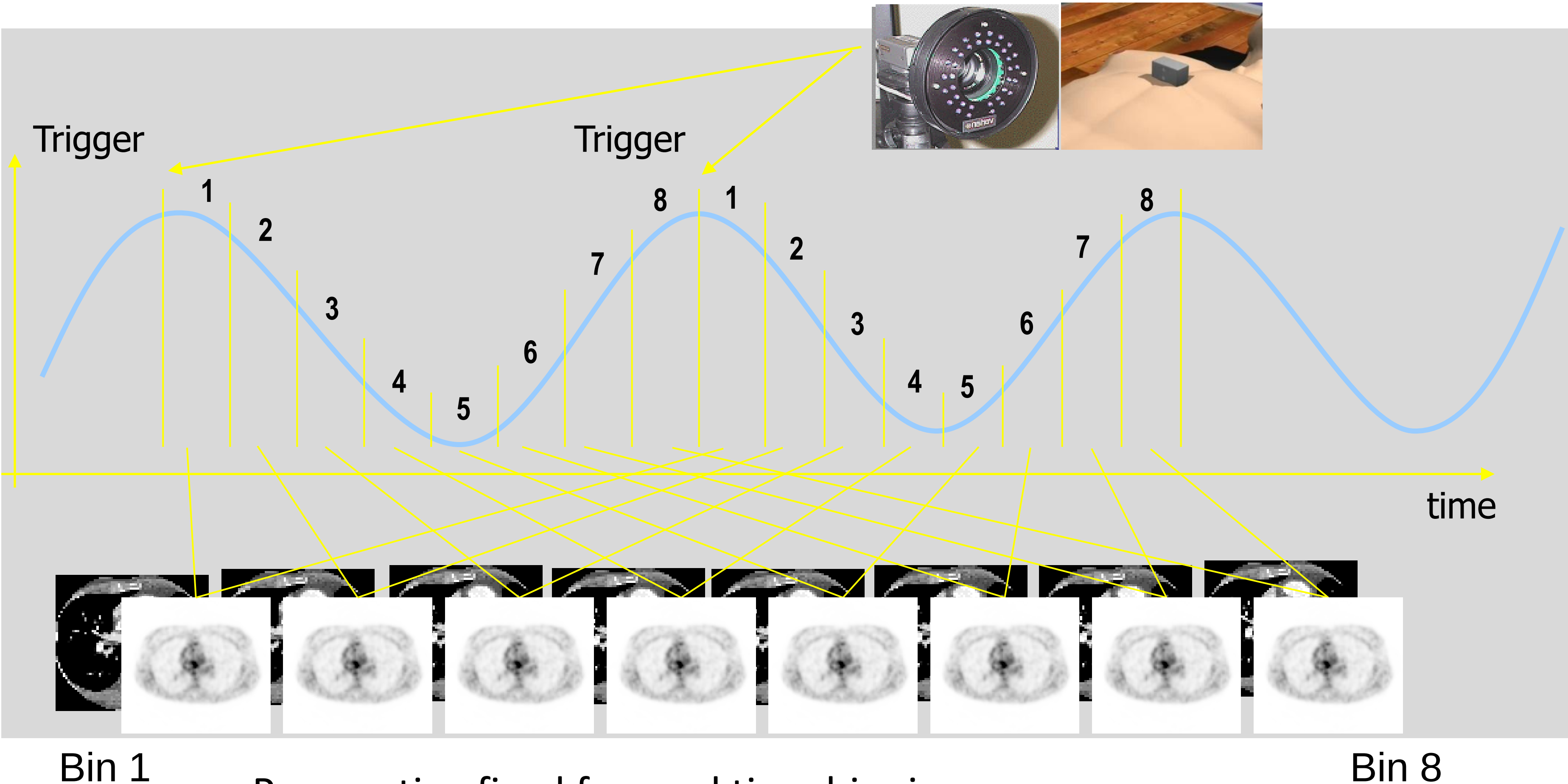

**SUV = 5.0**




**SUV = 8.5**

*Image courtesy: Tinsu Pan*

# Gated 4D PET and 4D CT Acquisition

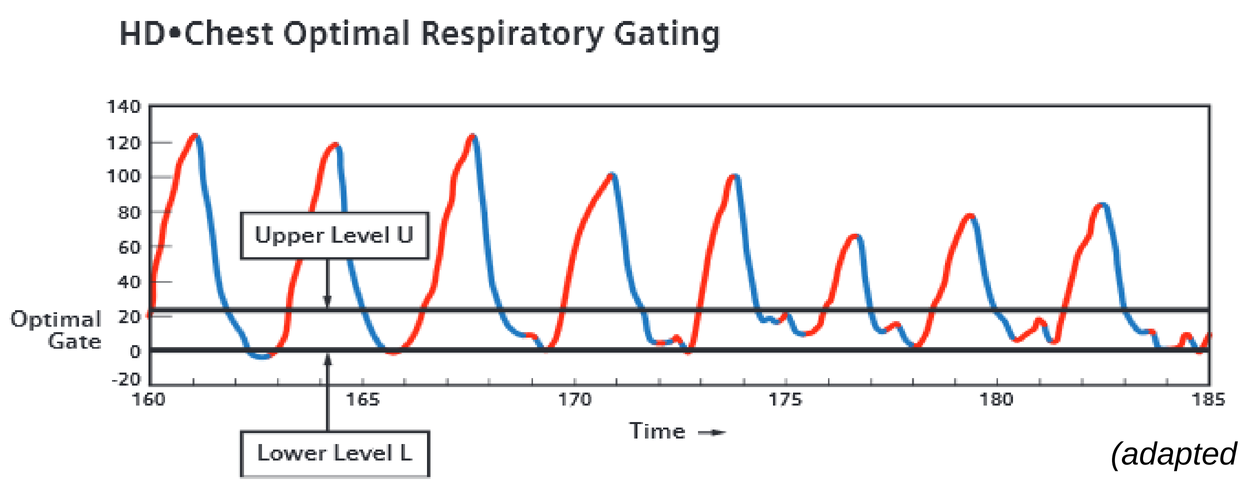



- Prospective fixed forward time binning
- Single FOV Gated PET and Gated CT
- User defined number of bins and bin duration
- Images will be noisy unless acquired for longer durations

*Image Courtesy: Tinsu Pan*

# Motion Correction Software

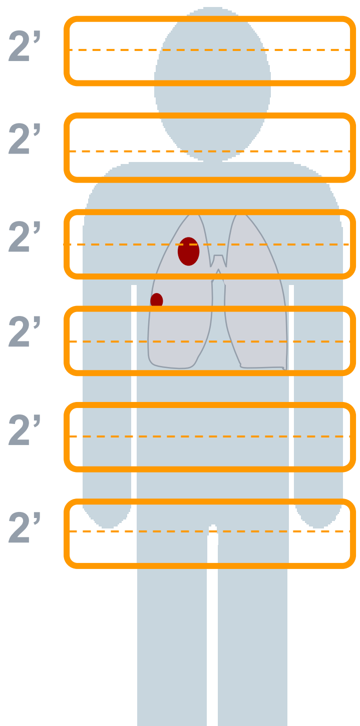
- Goal is to improve image quality, contrast, and quantitative accuracy – respiratory motion
- Q.Freeze (GE): Phase-matched 4D PET/CT
- Q.Static (GE) and HD.Chest (Siemens): Use PET data from end-expiration when motion is low
- Other vendors also have 4D PET solutions

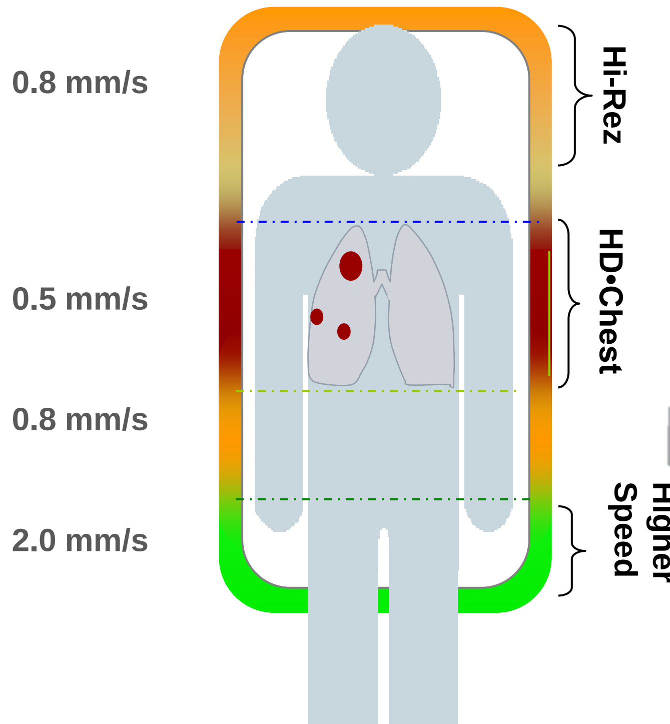



*(adapted from Siemens Healthcare)*

# Siemens Biograph mCT: FlowMotion

*Step-and-Shoot*



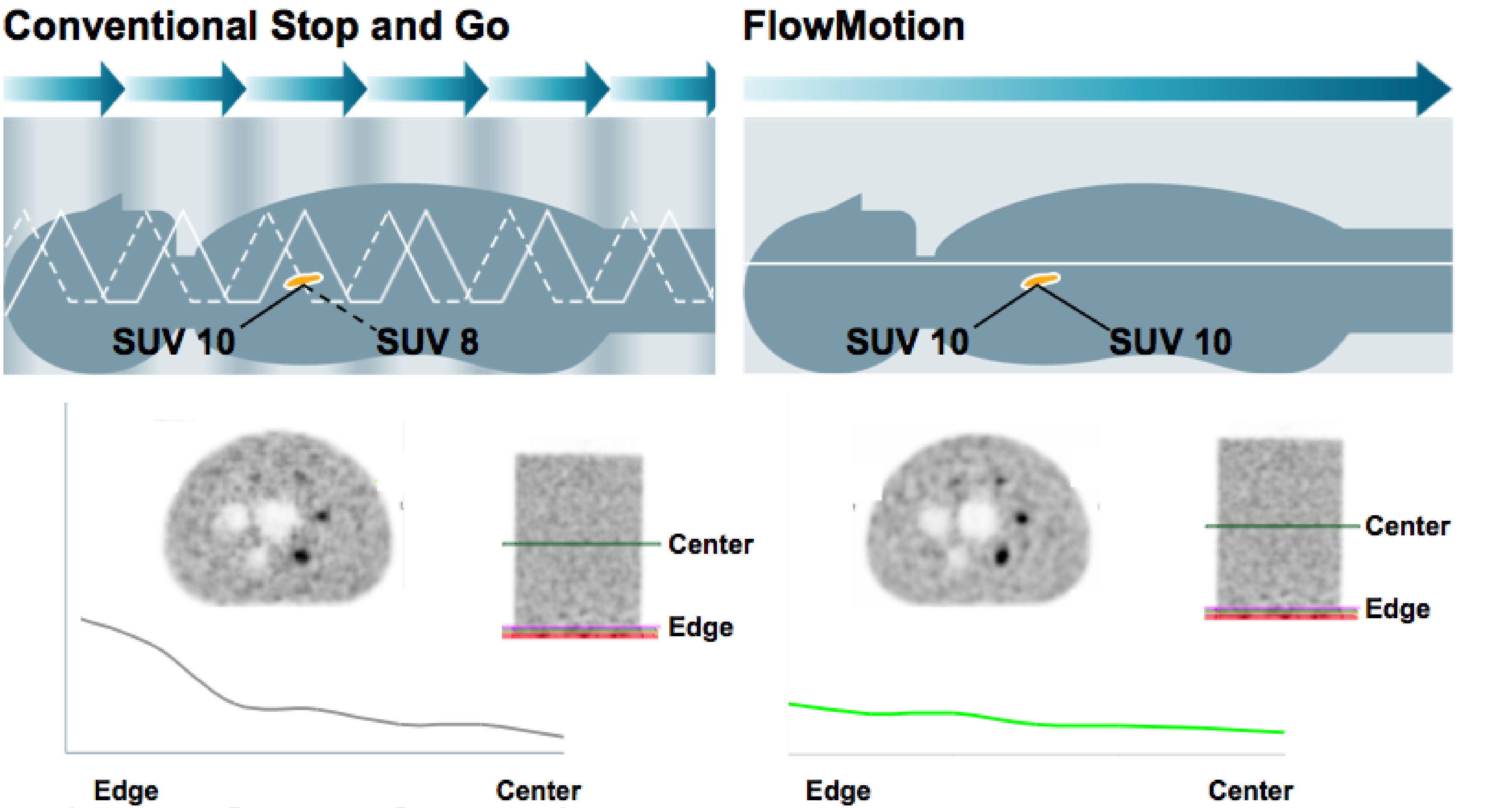





*(adapted from Siemens Healthcare)*

# Continuous Bed Motion

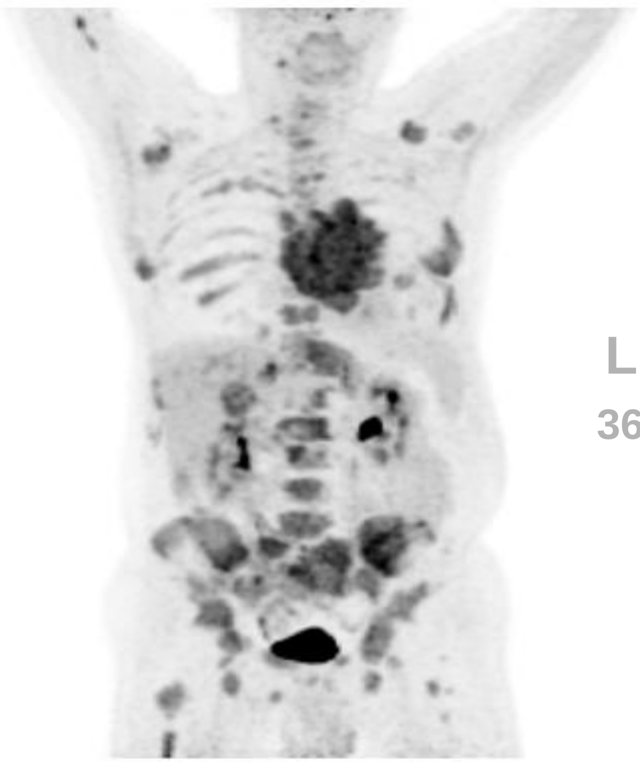
- Siemens FlowMotion mCT scanner




*(adapted from Siemens Healthcare)*

# Improved I.Q. – Reduced noise in end planes for every patient

**FlowMotion**

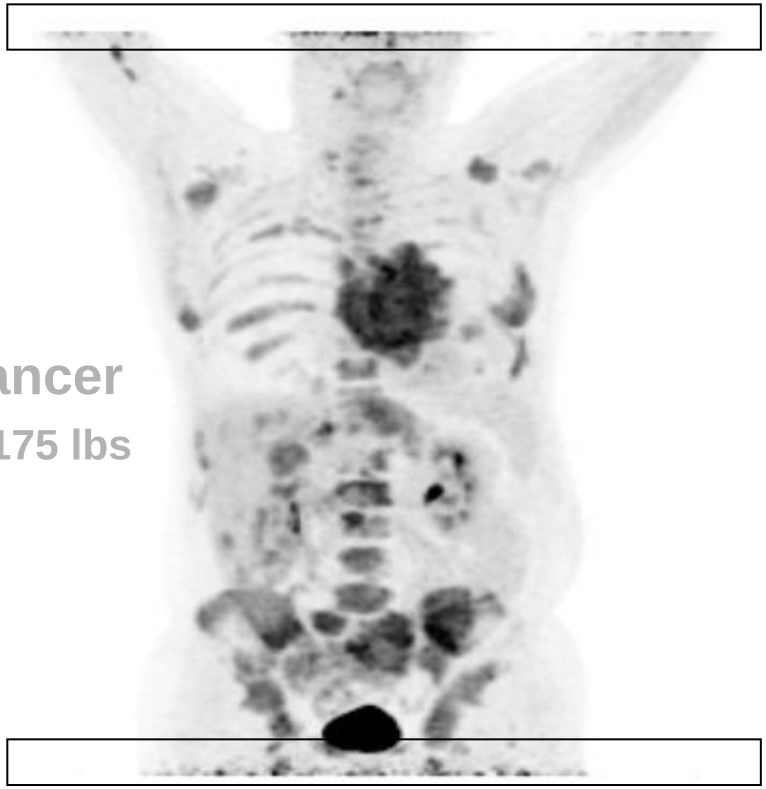


1.5 mm/sec
10 min Total Time
80 min P.I.

**Lung Cancer**

**366 MBq, 175 lbs**

**Step-and-Shoot**

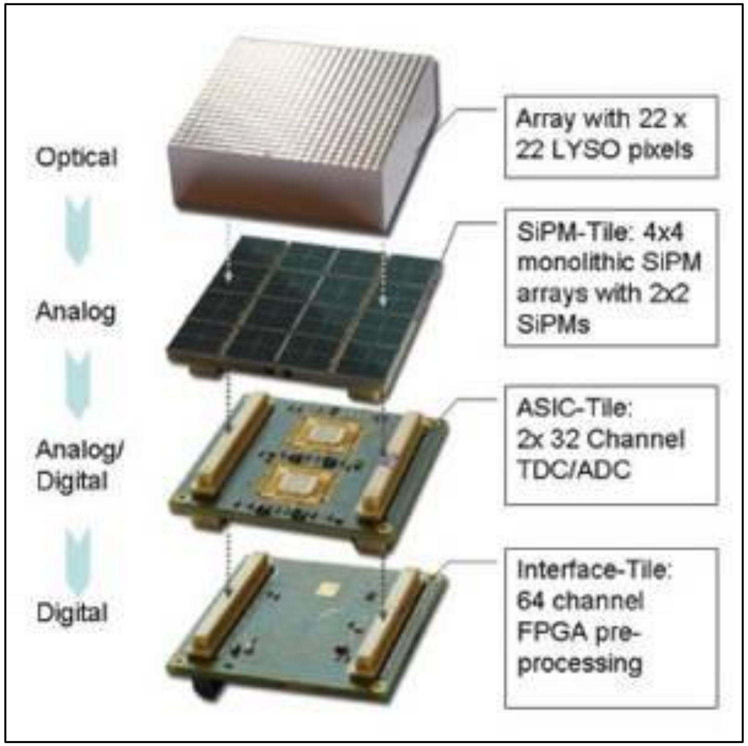


1.5 min/bed
15 min Total Time
60 min P.I.

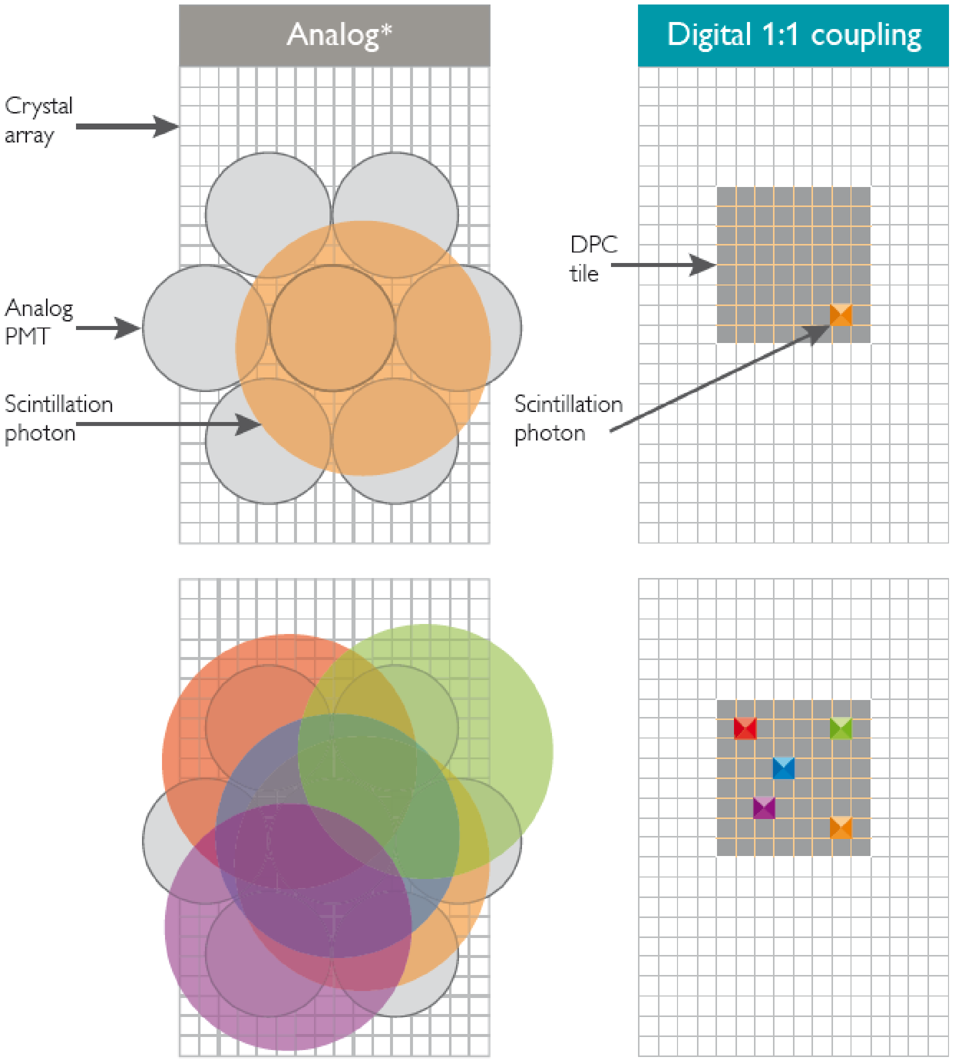
*(Image courtesy: UT Medical Center)*

# Fully Digital PET/CT – Philips Vereos

- LYSO crystals + SiPM → Fully digital detectors
  - Fast and high sensitivity
- TOF, PSF modeling, 4D capability

# SSPM – Digital photon counting




Improves resolution:
- No detector positioning

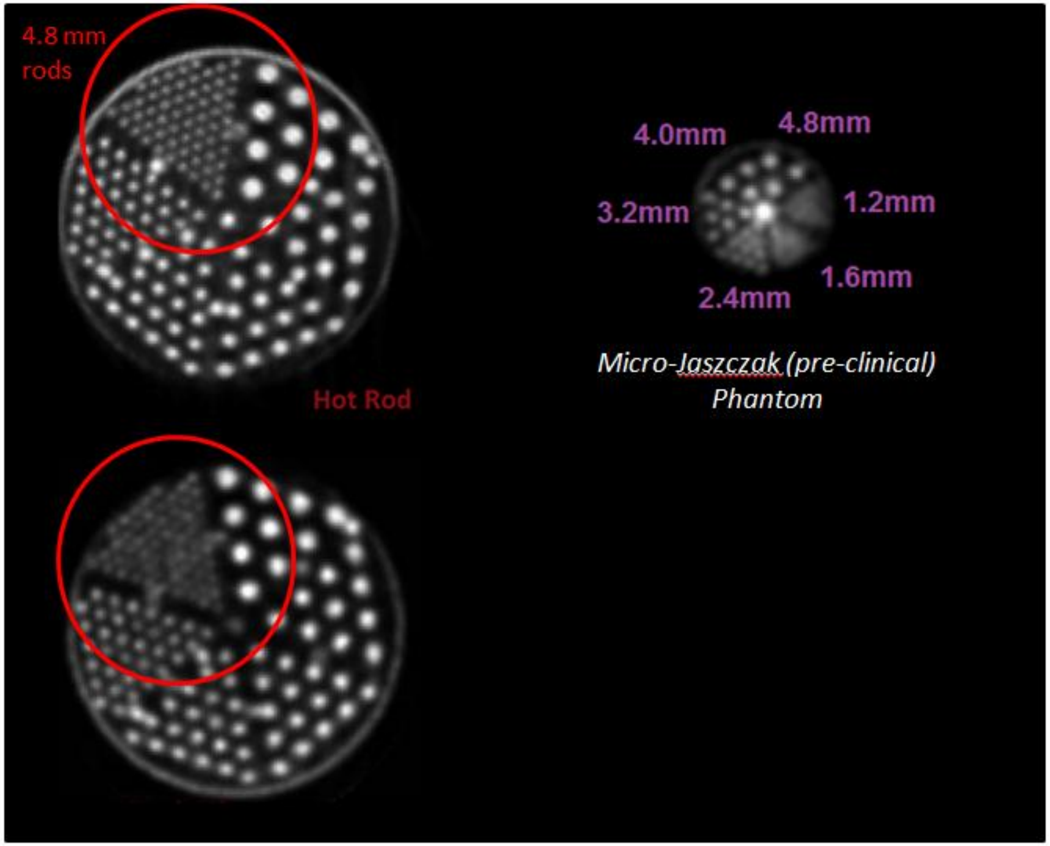
Improves sensitivity:
- high photon detection Eff.
- fast timing (high CNTR)
- improved TOF (~ 300 ps)
- decreased dead-time

Adapted from Philips Healthcare

## Improved spatial resolution seen with conventional clinical phantoms.

**Digital**

Analog*



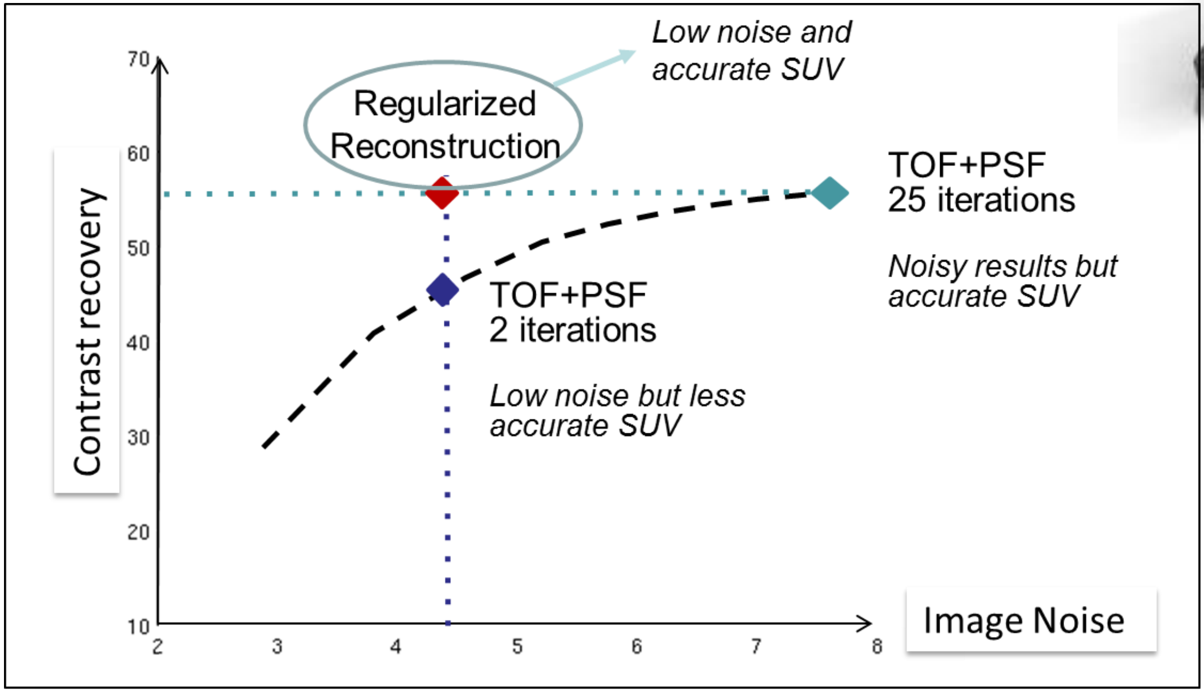

*Micro-Jaszczak (pre-clinical) Phantom*

*Deluxe Jaszczak Phantom*

**GEMINI TF 16*

# GE: Discovery IQ

- Regularized Reconstrution (Q.Clear)
- Achieve full convergence at lower image noise




(adapted from GE HealthCare)

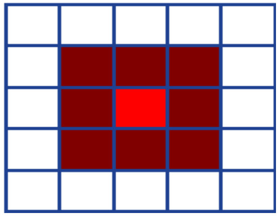
# Regularized Reconstruction Technology

Data statistics (likelihood)
General OSEM

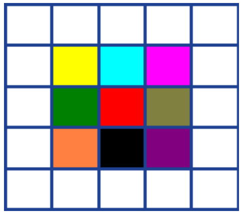
Regularization

$$\hat{x} = \arg\max_{x \geq 0} \sum_{i=1}^{n_d} y_i \log[Px]_i - [Px]_i - \beta \sum_{j=1}^{n_v} \sum_{k \in N_j} \kappa_{jk} \phi(x_j - x_k)$$

Weighting term to modulate the strength of the regularization term



Small pixel differences
Regularization favors this image

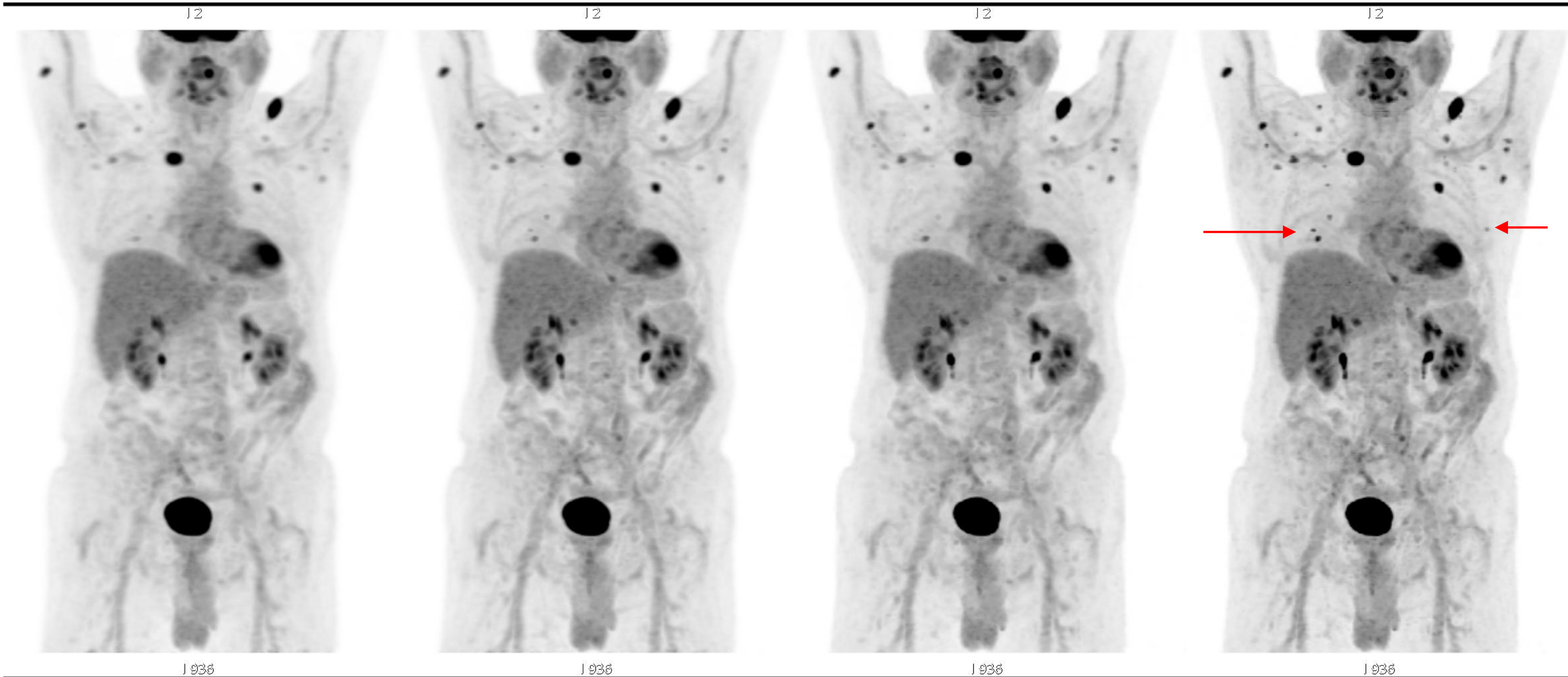


Big pixel differences
Regularization avoids this image

Adapted from GE HealthCare

# Regularized Reconstruction – GE Q.Clear

PSF TOF+PSF QC+PSF QC+TOF+PSF



77 years male with follicular lymphoma, 80 kg, 25 BMI, 9.4 mCi, 60 min post injection

# SAM Question 3

The well counter calibration for a PET scanner is used to:

| | |
|---|---|
| 0% | A. Correct for variations in image uniformity |
| 10% | B. Correct for variations in detector gains |
| 0% | C. Correct for differences in detector coincidence timing |
| 90% | D. Convert count rate (cps) to activity concentration (kBq/mL) |

# SAM Question 3: Answer

- The well counter calibration for a PET scanner is used to:

  A. Correct for variations in image uniformity
  B. Correct for variations in detector gains
  C. Correct for differences in detector coincidence timing
  D. Convert count rate (cps) to activity concentration (kBq/mL)

- **Answer: D**


- *Reference: SR Meikle, RD Badawi, "Quantitative Techniques in PET," in Positron Emission Tomography, eds. DL Bailey, DW Townsend, PE Valk, and MN Maisey, Springer-Verlag (London), 2005*

# SAM Question 4

The main advantage of a TOF PET scanner over a non-TOF PET scanner is:

- 50% A. Higher intrinsic spatial resolution
- 33% B. Higher image contrast-to-noise ratio (CNR)
- 12% C. Higher count-rate performance
- 5% D. Lower number of detector elements needed

# SAM Question 4: Answer

- The main advantage of a TOF PET scanner over a non-TOF PET scanner is:

  A. Higher intrinsic spatial resolution
  B. Higher image contrast-to-noise ratio (CNR)
  C. Higher count rate performance
  D. Lower number of detector elements needed

- **Answer: B**

- *Reference: M Conti, "Focus on time-of-flight PET: the benefits of improved time resolution," EJNMMI 38, 1147-1157, 2011*

# SAM Question 5

The minimum CT dose appropriate for PET/CT examinations are constrained by:

| | |
|---|---|
| 33% | A. Accuracy of CT-based attenuation correction |
| 60% | B. Radiologist preference for CT image quality |
| 0% | C. Equalize the CT dose to the PET dose |
| 7% | D. Accuracy of PET scatter correction |

# SAM Question 5: Answer

- The minimum CT dose appropriate for PET/CT examinations are constrained by:

  A. Accuracy of CT-based attenuation correction
  B. Radiologist preference for CT image quality
  C. Equalize the CT dose to the PET dose
  D. Accuracy of PET scatter correction

- **Answer: B**

- *Reference: FH Fahey, MR Palmer, KJ Strauss, RE Zimmerman, RD Badawi, ST Treves, "Dosimetry and adequacy of CT-based attenuation correction for pediatric PET: Phantom study," Radiology 243, 96–104, 2007*