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 = \( 100 \times \text{FWHM}/E \propto 1/\sqrt{E} \)
Collimators

NaI Crystal

Absorptive Collimation

$\gamma$ source
Collimators

NaI Crystal

Absorptive Collimation

$\gamma$ source
Collimator Resolution

\[ R_g = \frac{D(L_e + H + B)}{L_e} \]

System Resolution

\[ R_s^2 = R_i^2 + R_g^2 \]

Cherry, Sorenson, & Phelps, Physics of Nuclear Medicine, 2003
Collimator Efficiency

\[ G = \theta F \text{ where } \theta = C \left( \frac{D}{L_e} \right)^2 \]

\( \theta = \text{fraction of } 4\pi \)

\( F = \text{exposed fraction} \)

Parallel Hexagonal hole \( C = \frac{3}{8\pi} \)

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

**LEHR = 1.3 \times 10^{-4}**

**MELP = 3.1 \times 10^{-4}**

<table>
<thead>
<tr>
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<td>Isotope</td>
<td>99mTc</td>
<td>99mTc</td>
<td>99mTc</td>
<td>99mTc</td>
<td>99mTc</td>
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<td>Hole Shape</td>
<td>Hex</td>
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<td>Hex</td>
<td>Hex</td>
<td>Hex</td>
<td>Hex</td>
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<td>Number of Holes (x1000)</td>
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<td></td>
<td></td>
<td></td>
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<tr>
<td>Hole Length (mm)</td>
<td>24.05</td>
<td>24.05</td>
<td>24.05</td>
<td>24.05</td>
<td>24.05</td>
<td>24.05</td>
<td>24.05</td>
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<tr>
<td>Septal Thickness (mm)</td>
<td>0.16</td>
<td>0.13</td>
<td>0.13</td>
<td>0.13</td>
<td>0.13</td>
<td>0.13</td>
<td>0.13</td>
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<td>Hole Diameter (mm across the flats)</td>
<td>1.11</td>
<td>1.16</td>
<td>1.16</td>
<td>1.53</td>
<td>2.94</td>
<td>4</td>
<td>2</td>
<td>2.5</td>
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<td>Sensitivity @ 10 cm (cpm/cm²)</td>
<td>202</td>
<td>100</td>
<td>100</td>
<td>280</td>
<td>310</td>
<td>147</td>
<td>147</td>
<td>185</td>
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<td>Geometric Resolution @ 10 cm (mm)</td>
<td>14.6</td>
<td>6.3</td>
<td>6.3</td>
<td>6.3</td>
<td>10.8</td>
<td>13.2</td>
<td>13.2</td>
<td>10.6</td>
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<td>System Resolution @ 10 cm (mm)</td>
<td>15.6</td>
<td>7.4</td>
<td>7.4</td>
<td>7.3</td>
<td>12.5</td>
<td>13.4</td>
<td>13.4</td>
<td>19.0</td>
</tr>
<tr>
<td>Septal Penetration (%)</td>
<td>1.5</td>
<td>1.5</td>
<td>0.8</td>
<td>1.0</td>
<td>1.2</td>
<td>3.5</td>
<td>3.5</td>
<td>3.4</td>
</tr>
<tr>
<td>Focal Length @ Exit Surface (mm)</td>
<td>n.a.</td>
<td>n.a.</td>
<td>n.a.</td>
<td>n.a.</td>
<td>n.a.</td>
<td>n.a.</td>
<td>n.a.</td>
<td>n.a.</td>
</tr>
<tr>
<td>Weight (lb)</td>
<td>42</td>
<td>45</td>
<td>56</td>
<td>67</td>
<td>136</td>
<td>296</td>
<td>296</td>
<td>260</td>
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<tr>
<td>Weight (Kg)</td>
<td>18.9</td>
<td>20.4</td>
<td>25.2</td>
<td>30.5</td>
<td>61.8</td>
<td>134.5</td>
<td>134.5</td>
<td>117.0</td>
</tr>
</tbody>
</table>

1. Values measured in accordance with NEMA Standards Publication NU-1 2001 using 39° 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**
  \( S \propto z^{-2} \times z^2 \sim \text{constant} \)

\[ z = L_e + H + B \]
Anger Logic for Event Position

- Interaction location based on relative signal between $X^+$ and $X^-$ (for $X$ location) & $Y^+$ and $Y^-$ (for $Y$ location)
  - $X = (X^+ - X^-)/(X^+ + X^-) \rightarrow$ range -1 to +1
  - $Y = (Y^+ - Y^-)/(Y^+ + Y^-) \rightarrow$ range -1 to +1

- Interaction Energy $\propto$ Total Signal = $X^+ + X^- + Y^+ + Y^-$
SPECT Acquisitions

SPECT acquires 2D projections of a 3D volume

Wernick & Aarsvold, Emission Tomography, 2004

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

© Yale School of Medicine
SPECT data corrections

- Measured Projections
- INT Uniformity Correction
- EXT Uniformity Correction
- MHR/COR
- Inter-frame decay
- Scatter Correction
- CT Attenuation Correction
- Collimator Resolution Modelling

SPECT in kBq/mL
Scanning Calibration

SPECT in counts/mL

FBP/IR reconstruction
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, $C_i(x,y,\theta)$
  $$S(x,y,\theta) = \sum_i k_i \times C_i(x,y,\theta)$$

- Subtract scatter prior to reconstruction
  $$P_{corr}(x,y,\theta) \rightarrow P(x,y,\theta) - S(x,y,\theta)$$

- Incorporate scatter into forward projection
  $$P(x,y,\theta) \rightarrow P_{corr}(x,y,\theta) + S(x,y,\theta)$$
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
- Use a measured attenuation map along with models of scatter and camera resolution to perform a far more accurate reconstruction

$$a_{i,j,k} = a_{i,j,k}^{AC} \times a_{i,j,k}^{collimator} \times a_i^{efficiency}$$
SPECT Iterative Reconstruction

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

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

```
Forward Projection

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

Back Projection

\[ \lambda(b) = \sum_{d=1}^{D} y(d) p(b, d) \]
```
Iterative Reconstruction Flow Diagram

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°
FBP 180°
FBP noise
IR noise
HU-to-μ (CT-AC) Transforms

LaCroix et al., IEEE TNS 41, 1994

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

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

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

- 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

CT

Smooth, re-bin CT to match SPECT
Register CT w/ SPECT

CT\textsubscript{AC}

Apply bi-linear transform on pixel-by-pixel basis

\(\mu\)-map

Other factors:
- SPECT projections
- Scatter estimates
- Collimator response

Reconstructed SPECT

Transition Matrix
\(a_{ijk}\)
Siemens – Symbia Intevo

Symbia Intevo
Base system highlights

Intuitive Hand Controller
Easy-to-use with descriptive controls

HD Detectors
High-definition digital detectors that provide energy-independent performance

Patient Positioning Monitor
Self-guided touch screen user interface with intuitive icons

Autocontour
Infrared body contour system that minimizes patient-to-detector distance

Internal Electrocardiogram
Fully integrated ECG in system bed for fast patient setup and less cumbersome cables

Diagnostic Spiral CT
2-, 6- and 16-slice CT configurations

Open Gantry
Patient-friendly integrated gantry design with 70 cm (27.5 in) opening for greater patient comfort

Detector Tilt
Wide variety of detector configurations adjustable to any study and patient type (e.g., gurney imaging, 76° cardiac)

Innovative Bed Design
Low patient bed for easy access with ergonomic patient comfort accessories

Diagnostic CT
Quantitative SPECT
Advanced SPECT/CT reconstruction

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Conventional SPECT/CT Technology

Mechanical fusion of SPECT and CT

SPECT

SPECT counts

Attenuation Map

CT reconstruction

CT

Recon

3D OS-EM (Flash3D)
See the Unseen
Differentiation of tissue boundaries in bone imaging

SPECT

SPECT counts

CT

Zone Map

Voxel-by-Voxel Reconstruction

xSPECT

3D OS-CGM w/ CT-based Zones

Unrestricted © Siemens AG 2014
See the Unseen
xSPECT reconstruction shows better image quality

Conventional SPECT
- OSEM 3D iterative

xSPECT
- OSCGM 3D iterative
See the Unseen
Improved bone edge resolution for optimal visualization of vertebral metastases

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

Quantify the Difference
xSPECT improves visual and quantitative assessment

Conventional SPECT/CT

xSPECT

SUV 14.9

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

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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
- Absolute quantitation of tracer uptake with Q.Metrix

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 out: are clipped

Our Solution:
- Density inside the CT FOV is recovered
- Objects outside the CT FOV are stored

Wide Bore System
Advanced CT Technology

Key Features
- 16 slices × 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
Quantitative SPECT

Q.Suite – the path to absolute quantitation

- 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 performance**

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Axial field-of-view</td>
<td>14 cm in a single 360° rotation</td>
</tr>
<tr>
<td>Maximum rotation speed</td>
<td>12 seconds for 360° rotation</td>
</tr>
<tr>
<td>Maximum axial range</td>
<td>172 cm</td>
</tr>
<tr>
<td>Transaxial field-of-view</td>
<td>47 cm</td>
</tr>
<tr>
<td>Spatial resolution</td>
<td>&gt; 15 lp/cm @ 10% MTF</td>
</tr>
</tbody>
</table>

**XCT physical assembly**

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Type of detector</td>
<td>Digital amorphous silicon, columnar CsI scintillator</td>
</tr>
<tr>
<td>Detector size</td>
<td>30 cm x 40 cm</td>
</tr>
<tr>
<td>Detector pixel</td>
<td>0.2 mm x 0.2 mm</td>
</tr>
<tr>
<td>Number of elements</td>
<td>3,145,728</td>
</tr>
<tr>
<td>Generator output</td>
<td>10 kW, pulsed (2 msec. to continuous)</td>
</tr>
<tr>
<td>mA</td>
<td>5 – 80</td>
</tr>
<tr>
<td>kVp</td>
<td>120 rotating anode</td>
</tr>
</tbody>
</table>
High-Resolution CT images

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

Source: Images courtesy of Inselspital Bern University Hospital, Switzerland
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

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

Registration
Segmentation
2D/3D data
User Calibrations
Dose Calculation
Evaluation Tools
TAC, DVH, VOI stats
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 $\mu$-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
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
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
PET/CT
Annihilation Photons

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

\[ p = n + \beta^+ + \nu \; ; \; ^A\!X_Z = ^A\!Y_{Z-1} + \beta^+ + \nu \]

- Cyclotron (generator) for production of $\beta^+$ emitters

- $\beta^+$ annihilation $\rightarrow$ two simultaneous 511 keV photons
  - Emitted (nearly) 180 degrees apart

- Energy spectrum of $\beta^+$ emission is continuous
  - F18: $E_{\text{max}} = 0.64$ MeV, Range ~1 mm
  - Ru82: $E_{\text{max}} = 3.15$ MeV, Range ~1.7 mm
Schematic of a PET scanner

Nucleus

Detector Ring

positron

electron

Annihilation photon

Annihilation photon
# PET detectors

<table>
<thead>
<tr>
<th>Scintillator</th>
<th>Relative light output [NaI(Tl)=100]</th>
<th>Decay time (ns)</th>
<th>Thickness for 90% efficiency at 511 keV (cm)</th>
</tr>
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<tbody>
<tr>
<td>BGO</td>
<td>15</td>
<td>300</td>
<td>2.4</td>
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<tr>
<td>GSO</td>
<td>25</td>
<td>60</td>
<td>3.3</td>
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<tr>
<td>LSO, LYSO</td>
<td>80</td>
<td>40</td>
<td>2.7</td>
</tr>
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**PET Detector Block**
PET Detector Module and Rings

PET Detector Module

PET Detector Block

http://www.nucmed.buffalo.edu
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

- Measured Prompts
  - Correct Random
  - Normalize
  - Correct Deadtime loss
  - Correct Geometry
  - Correct Scatter
  - Correct Attenuation

- Correct Axial Sensitivity

- Scanner Calibration

- FBP/IR reconstruction

- PET image in kBq/mL

- Measured "True"
Model-based Scatter Estimation

**Idea**: To estimate the number of scattered coincidence along a specific LOR (LOR AB in figure)

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 \[
\frac{\text{Bq/mL}}{\text{Bq/mg}}
\]

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

\[
SUV_{\text{lean}} = \frac{\text{decay – corrected dose/ml of tumor}}{\text{injected dose/patient lean body mass in grams}}
\]
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

Two functions for CT as part of NM exams

- **AC**
  - Higher (Diagnostic)
  - Ultra-low (CT-AC only)

**CT Dose Requirement**

- **Anatomic Localization**
  - Loss of anatomic and morphologic information
  - Loss of PET accuracy from incorrect CT-AC

- **Higher** (Diagnostic)
- **Moderate**
- **Ultra-low** (CT-AC only)
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

<table>
<thead>
<tr>
<th>$\Delta t$ (ps)</th>
<th>$\Delta x$ (cm)</th>
</tr>
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<tbody>
<tr>
<td>600</td>
<td>9</td>
</tr>
<tr>
<td>100</td>
<td>1.5</td>
</tr>
<tr>
<td>33</td>
<td>0.5</td>
</tr>
</tbody>
</table>

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

$SNR_{TOF} \approx \sqrt{\frac{D_{obj}}{\Delta x} SNR_{non-TOF}}$
TOF PET has higher Image Contrast

TOF PET

SUVmax=3.3

Non-TOF PET

SUVmax=1.9

Image Courtesy: Osama Mawlawi
**PSF Resolution Modeling**

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

*Lee et al., PMB 49, 2004*

*Pecking et al., Clin. Exp. Metastasis 29, 2012*
PET Image Quality w/ PSF modelling

WITH

Max SUV = 8.4
Mean SUV = 1.9 (SD = 0.3)

WITH

Max SUV = 16.4
Mean SUV = 1.9 (SD = 0.3)

Max SUV = 19.4
Mean SUV = 2.2 (SD = 0.4)

Max SUV = 25.5
Mean SUV = 2.2 (SD = 0.4)

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 ↑

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**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

\[ \text{SUV} = 5.0 \quad \text{SUV} = 8.5 \]

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

HD•Chest Optimal Respiratory Gating

(adapted from Siemens Healthcare)
Siemens Biograph mCT: FlowMotion

*Step-and-Shoot*

0.8 mm/s

0.5 mm/s

0.8 mm/s

2.0 mm/s

(adapted from Siemens Healthcare)

S. Cheenu Kappadath, PhD
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.

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 $\rightarrow$ 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.
GE: Discovery IQ

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

(adapted from GE HealthCare)
Regularized Reconstruction Technology

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

Data statistics (likelihood)

General OSEM

Regularization

Weighting term to modulate the strength of the regularization term

Adapted from GE HealthCare
Regularized Reconstruction – GE Q.Clear

<table>
<thead>
<tr>
<th>PSF</th>
<th>TOF+PSF</th>
<th>QC+PSF</th>
<th>QC+TOF+PSF</th>
</tr>
</thead>
</table>

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)
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

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

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

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