An introduction to PET instrumentation

Tom K. Lewellen
Division of Nuclear Medicine
University of Washington Medical Center
Seattle, Washington USA
PET - the basic steps to make an image

1. Primary Detection
2. Decoding
3. Detector corrections
4. Coin. processing
5. Data binning
6. Data corrections
7. Image reconstruction
Proton decays to neutron in nucleus - positron and anti-neutrino emitted.

Unstable parent nucleus

Positron combines with electron and annihilates

Two anti-parallel $511\text{ keV}$ photons produced

PET basics - positron annihilation
What do we use for detectors?
Some details about scintillators

Ideally - want dense, fast, bright, cheap

So far, we can not get all at once
BGO versus NaI(Tl) - a simplistic comparison

- Dedicated BGO Ring system
- Dual head Coincidence Gamma camera

No energy discrimination

- 3 cm thick bismuth germanate oxide - BGO
  - $\mu = 0.91 \text{ cm}^{-1}$
  - Interaction probability = 0.93
  - Coincidence probability = 0.87

- 1.6 cm thick Sodium Iodide - NaI(Tl)
  - $\mu = 0.34 \text{ cm}^{-1}$
  - Interaction probability = 0.42
  - Coincidence probability = 0.18
Two basic detector design approaches

• Driven by scintillator characteristics
  • Pixilated detectors - discrete crystals or saw cut “slabs”
  • Continuous detectors - no longer a significant player in dedicated PET
Large pixilated detectors - evolution from large NaI based PET systems.

Current system (Philips) uses 4x6x20 mm GSO crystals mounted to a light guide and the array of PMTs - decoding is done essentially the same way as was done for the NaI version.
An example of a BGO Detector Block

Alternative to saw cuts - discrete crystals with surface treatments/coupling compound difference to produce equivalent light sharing.
### Block Crystal Decoding

**A, B, C, D = PMTs**

Example for a 6x6 array of crystals

<p>| | | | |</p>
<table>
<thead>
<tr>
<th></th>
<th></th>
<th></th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td>A</td>
<td>B</td>
<td>C</td>
<td>D</td>
</tr>
</tbody>
</table>

\[
E = A + B + C + D
\]

\[
Axial = Z = (C + D)/E
\]

\[
Transverse = X = (B + D)/E
\]
Example Crystal Map
Block Energy

Peaks for different crystals at different positions

Window center and width adjusted for each crystal
PET basics - coincidence detection

“Electronic” collimation => high sensitivity
Data is binned into a sinogram

Activity Distribution (single slice)

Data Acquisition

Image Reconstruction

Projection Data “Sinogram”

“distance”, d
Types of coincidence events

True

Scatter

Randoms
Randoms and count rate

Coincidence detection

- Randoms
- Prompt window
- Delayed window

Number of events vs. time between detector response.
Randoms and count rate - II

- We can also measure the detector singles rates
  
  \[ R = 2tS_1S_2 \]  Where \( t \) = resolving time, \( S \) is the detector singles rates

- Need to some other corrections - particularly accounting for energy windowing and proper dead time corrections.

- Pro = less noise in the correction.

- Con = is getting the corrections right and accounting for any rapid changes in counting rate during acquisition (works best in list mode if the activity is rapidly changing during the scan)
Scintillator speed and randoms

Randoms ~ constant versus time

In this example, #trues = constant

“fast” has ~ 1/2 the randoms
Randoms go as activity squared!

For full ring systems using BGO, LSO, or GSO randoms limit the maximum usable count rate.

Count Rate 3D (LLD = 375keV)

Activity Conc. (uCi/cc)
So, we can correct for randoms and we have a handle on count rate limitations.

Another major effect is that of photon attenuation in the patient.

How do we correct for that?
Options in collecting and processing transmission scan data

Direct coincidence measurement with rotating source

Singles measurements

Combined PET/CT scanner
Transmission scanning in PET

- Transmission rod sources rotate around the patient
- Acquisition is a high energy CT scan
- Data allows accurate attenuation correction
Effects of Attenuation in PET

\[ P_1 = e^{-\int_0^{x'} \mu(x)\,dx} \]

\[ P_2 = e^{-\int_{x'}^a \mu(x)\,dx} \]

\[ P_C = P_1 P_2 = e^{-\int_0^a \mu(x)\,dx} \]
Gated coincidence transmission

Coincidence + gating rejects most scatter

Problem - count rate limited by detector closest to rod source

Singles transmission

Count rate limited by the far detector

Problems - scatter; hardware modifications to tomograph

Solution - use image segmentation on attenuation data sets

Oct 2006
T+E - Transmission + emission
(transmission after dose is injected)

Essential for clinical scanning with conventional PET scanner

standard technique (without CT scanner)

Collect emission data

Collect transmission data

Use emission data to subtract emission contamination from transmission data

Either interleave T and E data or take all E data first and then T data
Segmented Attenuation Correction (SAC)

Smooth and reconstruct

Extract clusters

Histogram $\mu$ values

Classify image

Forward project
PET/CT

Images courtesy of CTI and GEMS web sites.

Oct 2006
PET/CT scan protocol

1. Scout scan (5-20 sec)
2. Selection of scan region (1-2 min)
3. Helical CT (1-2 min)
4. Whole-body PET (6-40 min)
The mass-attenuation coefficient ($\mu/\rho$) is remarkably similar for all non-bone materials since Compton scatter dominates for these materials. Bone has a higher photoelectric absorption cross-section due to presence of calcium.
CT-based Attenuation Correction

Bi-linear scaling methods apply different scale factors for bone and non-bone materials.
The presence of iodine confounds the scaling process as iodine cannot be differentiated from bone by CT number alone.
What we have so far:

- Basics of positron decay
- The kinds of coincidence events
- An overview of randoms and count rate issues
- Several options for attenuation correction

What are the options in basic tomograph designs?
Three basic types of coincidence imaging systems:

- Dual Head Coincidence
- “Low” cost dedicated PET
- “High” end Dedicated PET

- Lower cost
- Higher sensitivity and count rate
2D collimation - direct and cross planes

HR Mode

HS Mode
The basic limitation in image quality

The basic problem is getting enough Good Counts

To increase counts, what can you do?
• Increase patient dose
• Increase solid angle
  no collimation (3D)
  smaller detector ring diameter
  longer axial FOV
• Use more efficient scintillators
• Use more of the energy spectrum
What is the difference between “2D” and “3D” modes of operation
Impact of removing axial collimation and extending axial FOV

3. 3D provides increased sensitivity, but it is axially nonuniform.
3D body scans require overlapping FOVs

For a whole body scan, need more bed/detector positions as compared to 2D, but need less time per position.
3D PET - Single Events

Higher randoms and deadtime
An example of scatter fraction with no collimation:

- 2D 20 min. scan
- 3D 5 min. scan, no scatter correction
- 3D 5 min. scan, scatter correction
- 3D 20 min. scan, scatter correction
A quick review of a few points:

What we need are more good counts

BGO and LSO provide high sensitivity

3D mode offers more counts, but at the cost of higher randoms, dead time and scatter

Excellent for heads.
Still a matter of debate for whole body scans.

Let's take a brief look at image reconstruction
Another tool that is being developed to improve image quality in large patients is time-of-flight

- Measure time difference of detection of coincidence gammas
- Defines a line segment in space
- Result is an improvement in image signal to noise that is a function of the object size.
Why TOF?

• Increases effective SNR - shown by several groups

• As shown by the Penn group, TOF improves rate of convergence of statistical reconstructions with TOF resolutions on the order of 650 ps.
What Will Better Timing Resolution Bring?

Simulation: $2 \times 10^6$ Trues, $1 \times 10^6$ Randoms, Attenuation Included
OP-OSEM w/ TOF Extensions, 2 Iterations, 14 Subsets

*Data courtesy of Mike Casey, Siemens Medical Solutions*
What Limits Timing in Commercial PET? - slides from W. Moses

- Crystal Shape: 326 ps
- Light Sharing: 454 ps
- PMT Choice: 422 ps
- PMT Array: 274 ps
- Analog Electronics (CFD): 1354 ps
- Digital Electronics (TDC): 2000 ps

Commercial PET Timing Resolution (2003), CPS style block and electronics: 2.5 ns
Due to our limited time - we will focus on 2D image reconstruction.

Basic principles apply to 3D reconstructions

The big differences include:

How the data is binned for reconstruction

Options for converting the 3D data to 2D

How some corrections are applied
Image reconstruction - the basic flow

Load Raw Data

Correct for randoms (if not done real time)

Detector sensitivity corrections

Correct for radial distortions

Randoms

Normalization

Deadtime

Interpolation

Scatter

Attenuation

FBP or OSEM recon

Save Images
Uniform Cylinder (No Corrections)
Uniform Cylinder (With Atten)
Uniform Cylinder (With Atten and Norm)
Uniform Cylinder (With Atten, Norm, Sens)
Uniform Cylinder (With Atten, Norm, Sens, Act)
Detector failure
1. Estimate single scatters based on current emission & xmission images
2. Estimate multiple scatters with convolution model
3. Subtract scatter from emission sinograms and iterate
4. Scale final estimates to force mean value of pixels outside object to 0
Reconstruction of images from projections - FBP

Consider a pillbox of activity and a series of projections

angle = 0

angle = 90
Now we backproject the acquired data
To reduce the positive areas outside of the real object, we filter the projection data and then backproject. The filter to remove these radial artifacts

Apply a “window function” to trade off resolution and noise
What happens if we do not apply attenuation correction?

Non quantitative values and distortions.
What happens if we do apply attenuation correction?

Streaks due to noise amplification for the low count projections.
What we have just discussed is analytic reconstruction.

An alternative is a statistical reconstruction where you:

1) take into account the quality of the data
2) incorporate knowledge about the imaging system
A generic iterative procedure

- There are many ways to:
  - model the system (and the noise)
  - compare measured and estimated projection data
  - update the image estimate based on the differences between measured and estimated projection data
  - decide when to stop iterating
FORE = FOurier REbinning

Uses the Frequency-Distance Principle, which states that the signal at $(\omega, k)$ in the 2D Fourier Transform of a sinogram arises primarily from activity at a distance $r \propto -k/\omega$ from the center of the field of view.
With \( r \) determined by the Frequency-Distance Principle, the axial displacement of the data on a cross-plane sinogram at oblique angle \( \theta \) is calculated by \( \Delta z = r \tan \theta \).
FORE Algorithm Steps

1. Apply corrections
2. For each sinogram
   * take 2D FFT.
   * rebin sinogram according to Frequency Distance Principle
3. Normalize and take 2D IFFT of 2D sinograms.
4. 2D reconstruction of images
FORE allows fast 3D reconstruction using 2D techniques.

Maintains image quality and data integrity.

No compression or mashing.

Rebinning Results

Source at 0 10 20 cm

3D Recon
Slow, high image quality

SSRB
Fast, poor image quality

FORE
Fast, high image quality
A final thought

How many counts do we really need?

As many as we can get!!

It will be most interesting to see what technology gets us there!
Some things to note:

Dead time is primarily a function of the detector singles rate (pileup)

Randoms are a function of the timing discriminator output rate

If broad energy window(s) are used, the randoms rates will be higher

An aside - what is the singles counting rate?
A high total detector singles rate => pile-up

The results are:

- loss of events (dead time)
- misplaced events (loss of resolution)
**Where is tomograph design heading?**

- DOI detectors, fast scintillators, Small FOV, can use high cost per cc scintillator
- The main problem remains getting enough counts for the desired image resolution
  - Emphasis on PVI (3D) for high sensitivity and TOF
  - Thick (long) crystals for sensitivity
  - Coarse collimation to improve trues/singles?
  - DOI - improved axial sampling?
  - New scintillators only if cost is much lower.

**SPECT/PET systems, thick NaI or layered Crystals - not a major direction of MFRs.**
An VERY exaggerated example of the need for counts, Breast CA, 8.43 mCi FDG (primary tumor)

**VG CoDe - FBP**
- coronal
- transverse
- sagittal

30 min acq, 3.5 hour delay
2.3M counts
12mm Hanning

**Advance - FBP**
- coronal
- transverse
- sagittal

7 min acq/AFOV
3 AFOV
12mm Hanning
An exaggerated example of the need for counts, Breast CA, 8.43 mCi FDG (metastatic lesion)

VG CoDe - FBP

Advance - FBP

30 min acq, 3.5 hour delay
2.3M counts
12mm Hanning

7 min acq/AFOV, one hour delay
3 AFOV
12mm Hanning
An exaggerated example of the need for counts, Breast CA, 8.43 mCi FDG (metastatic lesion)

VG CoSem

transverse

coronal

sagittal

30 min acq, 3.5 hour delay
2.3M counts
12mm Hanning

Advance FBP

transverse

coronal

sagittal

7 min acq/AFOV, one hour delay
3 AFOV
12mm Hanning
An exaggerated example of the need for counts, Breast CA, 8.43 mCi FDG (metastatic lesion)

**VG CoSem**
- **Coronal**
- **Transverse**
- **Sagittal**

30 min acq, 3.5 hour delay
2.3M counts
8 mm Gaussian

**Advance OSEM**
- **Coronal**

7 min acq/AFOV, one hour delay
3 AFOV
10 mm Gaussian

Oct 2006
Additional material....
BGO vs. LSO and GSO - impact on NEC count rate
(3D NEMA PET phantom, NEC (1R))

BGO 41% SF
600 ns int

LSO 25% SF
0.5 R
150 ns int
0.95 sens

GSO 25% SF
0.75 R
200 ns int
0.8 sens
Count-rate Performance - C-PET

- current NEMA 20 x 19 cm cylinder
- Line source in 20 x 70 cm cylinder
- Phantom data compared to patient data

Graphs showing counts vs. activity concentration for trues, totals, and randoms.
Example of a Blank Scan
1. Estimate single scatters based on current emission & xmission images
2. Estimate multiple scatters with convolution model
3. Subtract scatter from emission sinograms and iterate
4. Scale final estimates to force mean value of pixels outside object to 0
FORE = FOurier REbinning

Uses the Frequency-Distance Principle, which states that the signal at \((\omega, k)\) in the 2D Fourier Transform of a sinogram arises primarily from activity at a distance \(r \propto -k/\omega\) from the center of the field of view.
3D-PET: The FORE Solution

With $r$ determined by the Frequency-Distance Principle, the axial displacement of the data on a cross-plane sinogram at oblique angle $\theta$ is calculated by $\Delta z = r \tan \theta$. 
1. Apply corrections
2. For each sinogram
   - take 2D FFT.
   - rebin sinogram according to Frequency Distance Principle
3. Normalize and take 2D IFFT of 2D sinograms.
4. 2D reconstruction of images
Rebinning Results

- FORE allows fast 3D reconstruction using 2D techniques.
- Maintains image quality and data integrity.
- No compression or mashing.

<table>
<thead>
<tr>
<th>Source at</th>
<th>0</th>
<th>10</th>
<th>20 cm</th>
</tr>
</thead>
<tbody>
<tr>
<td>3D Recon</td>
<td>-</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>Slow, high image quality</td>
<td>-</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>SSRB</td>
<td>-</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>Fast, poor image quality</td>
<td>-</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>FORE</td>
<td>-</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>Fast, high image quality</td>
<td>-</td>
<td>-</td>
<td>-</td>
</tr>
</tbody>
</table>
PET/CT scan protocol

Spiral CT
(1-5 min total)

Whole-body PET
(6-40 min total)

scatter correction
attenuation correction

Image fusion

Fused PET
T+E with a combined CT scanner

CT flux is so great that you can ignore the 511 keV gamma rays from the patient dose.

Current systems take the CT scan first (high speed spiral CT scanner).

Isolate bone and rescale the CT data for 511 keV.

Can not do interleaved E and T scans.

Several issues still to be resolved, for example:

- CT field-of-view smaller than PET FOV
- Respiratory motion

Dr Kinahan will cover this topic in detail in a later talk!