Positron Emission Tomography - PET

Positron Emission Tomography

Positron Emission Tomography (PET): Coincidence detection of annihilation radiation from positron-emitting isotopes followed by tomographic reconstruction of 3-D activity distribution.

Some unique features of PET:

- Use of “electronic collimation” instead of lead collimation.
- High detection efficiency
- Uniform resolution
- Accurate attenuation correction
- “Absolute” Quantification
- Use of short-lived biologically active radio-pharmaceuticals:
  - $^{11}$C-glucose
  - $^{13}$N-ammonia
  - $^{15}$O-water
  - $^{18}$FDG
  - $^{18}$FDOPA
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Coincidence Detection

- $^{11}_{5}$C
- $^{11}_{5}$B
- $\nu$
- $\beta^+$
- $e^-$
- Positronium
- 511 keV
- 180°
PET Gantry

ECAT HR+

PET Gantry

M. Dahlbom M284B Winter 2016
LSO 13x13 elements/block – 4x4x20mm² detector elements

Coincidence Detection
Positron Emission Tomography - PET

Coincidence Detection

The number of possible LORs:
\[ N_{\text{LOR}} = \frac{N(N-1)}{2} \]

Spatial Resolution

The spatial resolution in PET is primarily determined by:

- Detector size
- Physics of positron decay
- System geometry
- Detector material
Spatial Resolution

For a source placed at the midpoint between two scintillation detectors with a width $w_d$, the geometric line spread function has a triangular shape with a FWHM of $w_d/2$.

\[
\text{FWHM} = \frac{w_d}{2}
\]

Spatial Resolution - Tangential

For sources located between the midpoint and the detector surface the LSF will have a trapezoidal shape with width varying from $w_d/2$ (at the center) and $w_d$ at the detector surface.
Spatial Resolution - Radial

Parallax Error

Transaxial Resolution or ECAT EXACT HR

Correct line of response
Mis-positioned line of response

Small Lesion Detection: A Phantom Study

All spheres contain the same activity concentration

Bernard Bendriem, Ph.D.
Vice-President, R&D
March 19, 2004

Profile (10 mm)

Recovery coefficients

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Spatial Resolution

Although the most energetic positrons can travel several mm before annihilating, only a few of these are emitted.

The average positron energy emitted is approximately 1/3-1/2 of the maximum energy.

The total path length the positrons travel is not along a straight path. Through inelastic interactions with electrons in the positrons path is deflected. The distance from the mother nucleus is therefore much shorter.

From Levin & Hoffman PMB 44, 1999

Positron Range

From Levin & Hoffman PMB 44, 1999
Positron Emission Tomography - PET

Positron Range

\[ ^{18}\text{F} \]
635 keV

\[ ^{124}\text{I} \]
1.53 & 2.14 MeV

~65%

~50%

Non-colinearity

100 cm Ø ~ 2.5 mm FWHM

15 cm Ø ~ 0.3 mm FWHM

~0.3° FWHM
Spatial Resolution

The measured resolution (intrinsic resolution) of the system is a convolution of the various resolution components.

If the different resolution components are assumed to be Gaussian in shape and are described by a FWHM then the combined resolution is the squared sum of the individual resolution components:

\[
FWM_{\text{total}} = \sqrt{FWM_{\text{detector}}^2 + FWM_{\text{positron}}^2 + FWM_{\text{angulation}}^2}
\]
3D vs. 2D PET

The main advantage of the 3-D acquisition in PET is an improved sensitivity of ~5-7 times the 2-D sensitivity. The drawback is that the scatter fraction increases by a factor of 3. Non-uniform axial sensitivity Higher Randoms Rates → Increased Noise (offsets sensitivity gain) Dead-time problems when using slow detectors Image reconstruction is more complex More data
Coincidence Detection

Timing Resolution

Timing spectrum showing the PHA trigger time variation for a pair of BGO detectors in coincidence. The two peaks correspond to two separate measurements where an additional delay of 64 ns of the stop pulse for channel-to-time calibration.
Coincidence Detection

All coincidence detection systems have a finite time resolution

- BGO \( \sim 6 \) ns FWHM
- NaI \( \sim 4 \) ns FWHM
- GSO \( \sim 2 \) ns FWHM
- LSO \( \sim 0.5 \) ns FWHM
Random Coincidences

Because of the finite width of the logic pulses that are fed into the coincidence circuit, there is a probability for random or accidental coincidences between unrelated events.

\[ N_R = 2\tau N_1 N_2 \]

Where \( 2\tau \) is the coincidence window (or \( \tau \) is the width of the singles pulses).
Event Types

True Event

Scattered Event

Random Event

Multiple Event

Signal-to-Noise

True Coincidences
~ Activity
Good events!

\[ S/N \sim \frac{T}{\sqrt{T}} \]
Signal-to-Noise

Random Coincidences
  ~ Activity²
  Can be accurately corrected for
  Correction increases image noise
  Detector material dependent

\[ \frac{S}{N} \sim \frac{T}{\sqrt{T + 2R}} \]

Signal-to-Noise

Scattered Coincidences
  ~ Activity
  Reduces Image Contrast
  Requires correction
  Analytical estimation
  Correction increases image noise

\[ \frac{S}{N} \sim \frac{T}{\sqrt{T + S + 2R}} \]
Signal-to-Noise

Multiple Coincidences:
~ Activity^3
Never saved
Source of Dead time

Improvements in PET Image Quality

PET III 1975
ECAT II 1976
NeuroECAT 1978
ECAT 931 1985
ECAT EXACT HR+ 1994

Nal
BGO

CTI/Siemens
PET Detectors

Most modern PET systems use a different detector technology where a large number of scintillation crystals are coupled to a smaller number of PMTs. In the block detector, a matrix of cuts are made into a solid block of scintillator material to define the detector elements. The depth of the cuts are adjusted to direct the light to the PMTs. The light produced in each crystal, will produce a unique combination of signals in the PMTs, which will allow the detector to be identified.

The Technology: HiRez

<table>
<thead>
<tr>
<th>Standard Detector</th>
<th>HI-REZ Detector</th>
</tr>
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<tbody>
<tr>
<td>6.4 mm x 6.4 mm</td>
<td>4.0 mm x 4.0 mm</td>
</tr>
<tr>
<td>64 crystals/block</td>
<td>169 crystals/block</td>
</tr>
<tr>
<td>144 blocks/scanner</td>
<td>144 blocks/scanner</td>
</tr>
<tr>
<td>9216 crystals/scanner</td>
<td>24336 crystals/scanner</td>
</tr>
<tr>
<td>3.4 mm slice width</td>
<td>2 mm slice width</td>
</tr>
<tr>
<td>47 slices</td>
<td>81 slices</td>
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</table>
### Scintillator Materials

<table>
<thead>
<tr>
<th></th>
<th>Nal (TI)</th>
<th>BGO</th>
<th>GSO</th>
<th>LSO</th>
<th>LYSO</th>
<th>LaBr$_3$</th>
</tr>
</thead>
<tbody>
<tr>
<td>Density [g/ml]</td>
<td>3.67</td>
<td>7.13</td>
<td>6.71</td>
<td>7.35</td>
<td>7.1</td>
<td>5.29</td>
</tr>
<tr>
<td>$1/\mu$ [cm]</td>
<td>2.88</td>
<td>1.05</td>
<td>1.43</td>
<td>1.16</td>
<td>1.2</td>
<td>~2</td>
</tr>
<tr>
<td>Index of Refraction</td>
<td>1.85</td>
<td>2.15</td>
<td>1.85</td>
<td>1.82</td>
<td>1.81</td>
<td>1.9</td>
</tr>
<tr>
<td>Hygroscopic</td>
<td>Yes</td>
<td>No</td>
<td>No</td>
<td>No</td>
<td>No</td>
<td>Yes</td>
</tr>
<tr>
<td>Rugged</td>
<td>No</td>
<td>Yes</td>
<td>No</td>
<td>Yes</td>
<td>Yes</td>
<td>Yes</td>
</tr>
<tr>
<td>Peak Emission [nm]</td>
<td>410</td>
<td>480</td>
<td>430</td>
<td>420</td>
<td>420</td>
<td>380</td>
</tr>
<tr>
<td>Decay Constant [ns]</td>
<td>230</td>
<td>300</td>
<td>60</td>
<td>40</td>
<td>41</td>
<td>25</td>
</tr>
<tr>
<td>Light Output</td>
<td>100</td>
<td>15</td>
<td>35</td>
<td>75</td>
<td>75</td>
<td>&gt;100</td>
</tr>
<tr>
<td>Energy Resolution</td>
<td>7.8</td>
<td>20</td>
<td>8.9</td>
<td>&lt;9</td>
<td>11</td>
<td>7.5</td>
</tr>
</tbody>
</table>
Improvements in PET Image Quality

LSO
ECAT HRRT

- CTI/Siemens
Corrections in PET

In most nuclear medicine procedures, the goal is to produce an image in which the gray scale or count density is directly proportional to the regional isotope concentration. In order to achieve this in PET it is necessary to apply a number of corrections:

- Attenuation of photons in tissue
- Non-uniform response of detector elements
- Random coincidence events
- Detection of scattered events
- Loss of counts at high count rates - dead-time
- Isotope decay
- Absolute Calibration & cross calibration with other instruments

How accurate these corrections are will have a direct impact on the quantitative measurement.
In PET imaging of the brain, the shape of the head can be approximated with an ellipse. The dimensions of the fitted ellipse can be estimated by first reconstructing the data without attenuation correction. Then an ellipse is drawn onto the image from which the attenuation correction can be derived. The attenuation correction is the applied to the data and the image is reconstructed again.

This method can be fairly time consuming, especially on system producing a large number of transaxial slices.

$\text{Attenuation Correction} = e^{\mu D}$
Attenuation Correction

Blank Scan

Transmission Scan

Without Image Segmentation

With Image Segmentation
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The use of X-ray CT for Attenuation Correction of PET Data

Thomas Beyer, Paul E. Kinahan, David W. Townsend, and Donald Sasin
Department of Radiology, University of Pittsburgh, Pittsburgh, PA 15213-2582

Figure 1. Outline of proposed PET/CT scanner.
**Positron Emission Tomography - PET**

**PET/CT**

- GE
- Philips
- Siemens

- H.U. $\rightarrow \mu_{70\text{ keV}}$

- $x\ 0.495$

- $x\ 0.406$
Time-of-flight PET

\[ s = v \cdot t \]
\[ R + x = vt_1 \]
\[ R - x = vt_2 \]
\[ 2x = v(t_2 - t_1) \Rightarrow x = \frac{c\Delta t}{2} \]
Time-of-flight PET

For ideal detectors, TOF would eliminate the need for image reconstruction, since the measurement would allow each event to be accurately positioned in space.

All detectors have a finite time resolution, or uncertainty in timing. This translates to an uncertainty in positioning.

- BGO ~ 5 ns, 75 cm
- NaI ~ 1.5 ns, 22.5 cm
- CsF, LaBr₃ ~ 0.45 ns, 6.7 cm
- BaF₂, LSO, LYSO ~ 0.3 ns, 4.5 cm

Figure 1. Image elements contributing to a LOR, for conventional PET (left) and TOF PET (right).
Time-of-flight PET

Even with a finite time resolution, using the TOF information an improvement in signal-to-noise ratio (S/N) can be achieved:

$$SNR_{TOF} = \frac{D}{\Delta x} SNR_{conv.} = \sqrt{\frac{2D}{c\Delta t}} SNR_{conv.}$$

PHILIPS

Time-of-Flight vs. Conventional PET

Better information sent to reconstruction

More precise localization of annihilation event improves image quality
Time-of-flight PET - 1980’s

Problems with TOF in the 80’s
Poor detection efficiency of available scintillators

TOF Gain did not offset the poor efficiency

To improve the efficiency, large detector modules were used

A more significant gain in S/N could be achieved by using high resolution detectors and conventional detection methods (Phelps, Hoffman, Huang, 1982).

Time-of-flight PET - 2006

Scintillators:
CsF, BaF$_2$ $\rightarrow$ LSO, LYSO - fast, high light, and dense

Detectors/PMTs:
1:1 coupling $\rightarrow$ 100:1 crystal encoding - spatial resolution

Geometry:
2D (septa) $\rightarrow$ 3D with large axial FOV - sensitivity

Reconstruction:
Analytic (FBP) $\rightarrow$ iterative (list-mode) - system modeling

Electronics:
Accurate and stable
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Can we see TOF improvement?

<table>
<thead>
<tr>
<th>Time</th>
<th>Non TOF</th>
<th>TOF</th>
</tr>
</thead>
<tbody>
<tr>
<td>5 min</td>
<td><img src="non_TOF_5min.png" alt="Image" /></td>
<td><img src="TOF_5min.png" alt="Image" /></td>
</tr>
<tr>
<td>3 min</td>
<td><img src="non_TOF_3min.png" alt="Image" /></td>
<td><img src="TOF_3min.png" alt="Image" /></td>
</tr>
<tr>
<td>1 min</td>
<td><img src="non_TOF_1min.png" alt="Image" /></td>
<td><img src="TOF_1min.png" alt="Image" /></td>
</tr>
</tbody>
</table>

6-to-1 contrast; 35-cm phantom

Noticeable improvement with TOF with large size phantom

Gemini TF - patient study

- **Rectal carcinoma** metastases in mesentery and bilateral iliac chains
- **Patient details**: 114 kg; BMI = 38.1
- **Injection details**: 12 mCi; 2 hr post-inj
- **Scanning details**: 3min/bed

Lesion contrast (SUV) improves with TOF reconstruction
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ME Phelps et. al.

DH Silverman et. al.
Positron Emission Tomography - PET

DH Silverman et. al.

DH Silverman et. al.
Low Grade Brain Tumor

FDOPA Uptake Patterns

0.0 0.5 1.0 1.5 2.0 2.5 3.0

SUV

0 20 40 60 80 minutes

Tumor reaches maximum before striatum
Positron Emission Tomography - PET

Integrated PET/MRI System

Opportunities:
• direct and accurate registration of molecular PET signal with high resolution anatomy
  – Anatomically guided analysis of PET data
  – Improved quantification of PET data
  – Good soft tissue contrast, no additional radiation dose
• time correlation of PET and MRI or MRS signal
  – Interventional, therapeutic studies
  – Dual-labeled agents (\(^{64}\text{Cu}, \text{Gd}\))

Images courtesy Bernd Pichler

Positrons in Magnetic Field

A \(^{39}\text{Y} (E_{\text{max}} = 3.15 \text{ MeV})\)

B \(^{125}\text{I} (E_{\text{max}} = 2.14 \text{ MeV})\)

0 Tesla

10 Tesla
Positron Emission Tomography - PET

**MR Compatible PET System**

- Animal MR System
- PET Detectors
- Magnet
- Gradient Coils
- RF coil

**Simultaneous PET/MRI Imaging**


- 56 mm ring diameter
- 72 2x2x25 mm LSO scintillators

200 g Rat - \(^{18}\)F-FDG Brain Study
Challenges in Combining PET and MR imaging

PET Detectors affected by:
- Static magnetic field
- Rapidly changing gradient field
- Radiofrequency signals

MR affected by
- PET detectors and electronics

B. Pichler et. al., 2008
Solutions for combining PET-MR

MR-Compatible PET Detector Module

scintillator array  optical fiber bundle  PSAPD  preamplifiers
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PMT vs. APD/SiPM

PET Insert

preamplifiers  PSAPDs  optical fibers  scintillator ring
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MR phantom images: GE (left) and SE (right) sequences of a gadopentetate dimeglumine/H2O phantom (T1 = 250 ms) without PET insert (first row) and with PET insert unpowered (second row) and powered (third row).
Catana et.al., JNM 47 (12), 2006
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MR/PET brain scanner prototype (conceptual design)

- LSO array
- Crystals
- 3x3 APD array
- 9-Channel preamplifier ASIC board
- 9-Channel driver board
- Avalanche Photo Diodes (APD)
- Integrated Cooling Channels

1. PET detector assembly for mMR. 64 Lutetium Oxyorthosilicate (LSO) crystals form one block that transforms 511 keV gamma quanta into light flashes. Light events in the LSO crystals are detected by a 3x3 array of avalanche photo diodes (APD). 9-channel preamplifiers and driver boards as well as integrated water cooling completes each detector block. This detector assembly is characterized by its small size, can be designed free of magnetic components, and performs in strong magnetic fields. 56 such blocks form one detector ring. 8 rings form the PET detector assembly in the Biograph mMR that spans a longitudinal field of view (FOV) of 25.8 cm.
Test Setup

- Concentric MR and PET
- MR:
  - CP-TXRX-Head coil
  - Inner Diameter 27 cm
  - RF Shield ID 36 cm
- PET:
  - 512 LSO crystals in 2 modules
  - FOV 3.2 cm
  - Imaging by phantom rotation
- MR/PET phantom
  - 1.0 mm - 3.5 mm diameter holes
  - Filled with water and 1.25 g NiSO$_4$ / litre and about 50 MBq FDG
SUPPLEMENTAL FIGURE 1 – Diagram of the Biograph mMR, depicting how the PET detectors are located within the MR coils.

SUPPLEMENTAL FIGURE 2 – Photograph of the Biograph mMR system installation at the Institute of Medical Physics, University of Erlangen. Both imaging modalities – MR and PET – are fully integrated into one MR/PET hybrid system.
PET-MRI Attenuation Correction

The Biograph mMR is equipped with the Tim (Total imaging matrix) RF coil technology consisting of multiple integrated surface coils that cover the patient’s body from head to toe with up to 102 RF coil elements connected to 32 RF receivers. This multichannel phased array RF coil configuration enables parallel imaging and whole-body MR data acquisition with optimized signal-to-noise (SNR) performance. In simultaneous MR/PET hybrid imaging, the surface RF coils are located between the radioactivity emitting patient and the PET detectors. As a consequence, the RF coils should be designed to be as PET-transparent as possible.
FIGURE 7. mMR PET/MR (A) and Biograph PET/CT (B) fused views of whole-body 18F-fluoride scan of same patient. mMR (C) and Verio (D) T2-weighted coronal views of healthy volunteer.