Medical Imaging techniques
(SPECT and PET)

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Rectilinear scanner 1951
Simple scintillator counter
I-131 Thyroid
Planar imaging
Ink intensity is proportional
to measured photon count-rate

H. Anger 1958
First Gamma Camera
Tc-99m Pertechnetate
Brain scan of a patient with a glioma
Which devices

- Imaging devices
  - Gamma camera
  - SPECT/CT
  - PET/CT (PET/MRI)
Gamma Camera
- Emission imaging is based on scintillation detection
- Line of sight are defined by collimation (W or Pb collimator)
Gamma camera: Problems in event localization

A) Valid event (useful for correct localization on the imaged region)
B) Scatter on the crystal
C) Scatter on the patient
D) Septal penetration

Incorrect source localization
Event localization on the crystal surface

Centroid (baricentrum) of the PM response

\[
x = \frac{\sum_{i=1}^{n} S_i x_i}{\sum_{i=1}^{n} S_i}
\]

\[
y = \frac{\sum_{i=1}^{n} S_i y_i}{\sum_{i=1}^{n} S_i}
\]

Software positioning also account for non-linearity
In PM response

Pulse height analysis and energy determination

Discriminate gamma events
Scattered photons

Threshold level on signal height
\(\rightarrow\) Acceptance energy window
(photoelectric peak)

No scatter (or low scatter) is accepted for image formation
**Gamma camera: Performances**

**Intrinsic spatial resolution** (detector) 
→ (4mm)
- Impinging gamma energy
- Crystal thickness

**Collimator resolution** (parallel-hole) 
~10 mm

\[ R_{\text{coll}} = d \left( \frac{l+b}{l} \right) \]

- Resolution increases with photon energy
- Resolution decreases with crystal thickness
Gamma camera Sensitivity

Collimator Efficiency

\[ S_{\text{coll}} = \frac{\Omega}{4\pi} \frac{A_{\text{holes}}}{A_{\text{crystal}}} = \frac{\pi \left( \frac{d}{2} \right)^2}{l^2} \frac{A_{\text{holes}}}{A_{\text{crystal}}} \]

\( \Omega \): solid angle subtended by the collimator hole from the source
\( d \): collimator hole diameter
\( l \): collimator length (septa length)

Total Efficiency

\[ S_{\text{tot}} = \frac{\Omega}{4\pi} \frac{A_{\text{holes}}}{A_{\text{crystal}}} \left[ 1 - e^{-\mu d_{\text{crystal}}} \right] \]

Detection efficiency (crystal)

<table>
<thead>
<tr>
<th>Collimator Type</th>
<th>Recommended Max. Energy (keV)</th>
<th>Efficiency, g</th>
<th>Resolution ( R_{\text{coll}} ) (FWHM at 10 cm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Low-energy, high-resolution</td>
<td>150</td>
<td>1.84 \times 10^{-4}</td>
<td>7.4 mm</td>
</tr>
<tr>
<td>Low-energy, general-purpose</td>
<td>150</td>
<td>2.68 \times 10^{-4}</td>
<td>9.1 mm</td>
</tr>
<tr>
<td>Low-energy, high-sensitivity</td>
<td>150</td>
<td>5.74 \times 10^{-4}</td>
<td>13.2 mm</td>
</tr>
<tr>
<td>Medium-energy, high-sensitivity</td>
<td>400</td>
<td>1.72 \times 10^{-4}</td>
<td>13.4 mm</td>
</tr>
</tbody>
</table>
Tomographic reconstruction
Bases of tomography reconstruction in nuclear medicine

Projections in a simple 1-D detector case

Signal proportional to the summed activity along the line of response (assumption of no attenuation and scatter)

Head rotation around the imaged object

Many projections at different angles are obtained
1-D projections of a unknown activity distribution $f(x,y)$
N projections $p(r,\phi)$ ↔ N angular samples ↔ N sinogram lines in the $(r,\phi)$ space

1-D Projection data are arranged in a 2D $(r,\phi)$ **SINOGRAM** representation

Each row ($\phi$) in the sinogram displays the intensity profile measured in the corresponding projection
Object space
Activity distribution

\( f(x, y) \)

Emission Tomograph
Measured projections

Emission Tomograph
Measured projections

\( p(r, \phi) \)

Sinogram

Forward projection
Calculated projections
\( p(r, \phi) \)

Compare /converge?

Yes

Reconstructed image
\( f'(x, y) \)

No

Update image estimate

Iterative reconstruction workflow

Iterative reconstruction process

Measured data set

Expectation maximization cost function
(convergence criterion)

\[
\begin{align*}
f_i^{k+1} &= \frac{f_i^k}{\sum_{j=1}^{m} M_{i,j}} \times \sum_{j=1}^{m} M_{i,j} \frac{p_j}{\sum_{i=1}^{n} M_{i,j} f_i^k} \\
p_j &= \sum_{i=1}^{n} M_{i,j} f_i
\end{align*}
\]

Method to upgrade the image estimate

Account for:
- Detector and collimator response,
- Finite spatial resolution
- Attenuation
- Scattering

Animations from: Floris HP van Velden, PhD, EANM Milan 2012
Detection/Localisation
Contrast
Quantification
SPECT/CT
Signal intensity in SPECT voxels is proportional to the amount of activity contained.

But absolute quantification (Bq/mL) is very hard to achieve.

- **Line of response** are not straight cylinders but are diverging cones.
- **Tissue attenuation** results in depleted signal from deeper location in patient.
- **Scattering** in patient, detector and collimator results in event mislocalisation.

An exact detector collimator response is needed (septal penetration, PSF resolution recovery).
CT based attenuation correction in SPECT

SPECT
Rotating gamma camera heads
CT gantry

CT-AC is standard in modern hybrid SPECT/CT devices

Tissue attenuation can be derived from transmission data (CT)

CT-Based Attenuation map
For attenuation Correction of SPECT data
Effective energy of the beam can be defined:

120 kVp $\rightarrow$ 75 keV

Appropriate $\mu(x,y,E_\gamma)$ map need:

Energy scaling the attenuation coeff. from effective beam energy to the given radionuclide energy

CT map is in Hounsfield units obtained from a broad-energy Spectrum

Segmentation on CT:
- Air
- Bone
- Soft tissues

Continuous attenuation map $\mu(x,y,E_\gamma)$ from CT
Scattering in patient, detector and collimator results in event mislocalisation.

Scatter correction by Chang corrections $\rightarrow$ smaller attenuation coefficient
Only works in regions of uniform activity distribution and attenuation

Scatter correction by measured scatter component subtraction in projections

Scatter correction by energy discrimination in PHA

Scatter fraction in the selected energy window in unfavorable condition can be up to 40%
Scatter correction in SPECT
PHA based energy discrimination

Typical energy spectrum from $^{99m}$Tc source
(without attenuating/scattering media)

Energy spectrum in the presence of scattering
Patient (especially obese patients)

Scatter correction must be performed before Attenuation correction
to avoid amplification of scatter contribution
Attenuating map data can be integrated into the Image Matrix (M)
To account for the probability of scatter radiation in the source region (x,y) to produce a signal into a given detector element (p_j)

Collimator/detector response can also be integrated in M
- Monte Carlo simulations (gold standard)
- Measured data
PET/CT
Basic principle of PET

Radionuclides used in PET: $^{11}$C, $^{13}$N, $^{15}$O, $^{18}$F, $^{68}$Ga, $^{82}$Rb
PET detector design

Modules

Bloc

PET ring

Detector elements (scintillators)
Energy spectra in $\beta^+$ decay

**Important:** the range of $\beta^+$ in matter depends on the energy with direct consequence in spatial resolution.
β+ emission: Energy and range

\[ ^{18}\text{F} \quad \text{(}E_{\text{max}} = 635 \text{ keV)} \]

\[ ^{15}\text{O} \quad \text{(}E_{\text{max}} = 1720 \text{ keV)} \]

- 100,000 Events
- 10,000 Events
- 0.102 mm FWHM
- 1.03 mm FWTM
- 0.501 mm FWHM
- 4.14 mm FWTM
Coincidence detection: Electronic collimation

Camera: Detector ring with detector blocs

Coincidence are events couple of events
- occurring at the "same time" (~ns)
- having the right energy (~511 keV each)
Coincidence detection

Only photons detected in coincidence are considered to build the image.

Diagram:
- LOR (Line of Response)
- Annihilation
- Signals in Detector 1 and Detector 2 are shown with corresponding timestamps.
- Sum of signals D1 + D2
- Coincidence window: $\tau$
Spatial localization of annihilation events

2-511keV photon in coincidence $\rightarrow$ 1 Line (tube) Of Response (LOR)

- **Goal**
  - Recover the exact position of the annihilation event

- **Problem**
  - We have not information about the place along the LOR where the annihilation happened
Direct localization using the Time of Flight (TOF) information

Measured detection time in D1 and D2

\[ T_1 = \frac{D - x}{c} \quad T_2 = \frac{x}{c} \]

\[ \Delta T = T_2 - T_1 = \frac{x}{c} - \frac{D - x}{c} = \frac{2x - D}{c} \]

Uncertainty on time estimation →
Error in annihilation position estimation

\[ \delta T = \frac{\delta x}{c} \]
Direct localization using the Time of Flight (TOF) information

• **Source 1**
  - \( t\gamma_1 = \frac{D_1}{c} \rightarrow 1.4 \text{ ns} \)

• **Source 2**
  - \( t\gamma_2 = \frac{D_2}{c} \rightarrow 1.2 \text{ ns} \)

• Difference in time of flight \( \Delta t = 0.2 \text{ ns} = 200 \text{ ps} \)

\[ c = 3 \times 10^{11} \text{ mm/s} \]

\[ \Delta x = \frac{c \Delta t}{2} \Leftrightarrow \Delta t = \frac{2 \Delta x}{c} \]
Improved pelvic nodule visualization with Time-of-Flight
Body attenuation greatly reduces counts. As size increases, counts are reduced exponentially. ToF gain is greater for large patients as it partially compensates for the lower quality of large patients.
Dual modality PET-CT

- **Goal**: improve activity localization and implement attenuation correction (auto-registration of anatomic CT and functional PET)
Modeling of the PSF improves actual positioning of the LoRs.

PSF $\rightarrow$ uniform space resolution across the FOV.

Source: Siemens

Source: GE
Spatial resolution

\[ R_t = K_r \cdot \sqrt{R_i^2 + R_p^2 + R_a^2 + R_l^2} \]

- \( R_i \) is related to the detector width (w)
  - from \( w/2 \) (center) to \( w \) (detector), 2 – 4 mm
- \( R_p \) is related to the positron range
  - 0.2 mm for \( ^{18}\text{F} \) and 2.6 mm for \( ^{82}\text{Rb} \)
- \( R_a \) is related to the \( \gamma \) non-collinearity
  - \( \pm 0.25^\circ \) deviation from 180\(^\circ\)
  - 1.8 mm for a 80-cm PET scanners
- \( R_l \) is related to the localization of detector
  - (use of block detectors instead of single detectors)
    - 2.2 mm for BGO (less for LSO)
- \( K_r \) is a factor related to the reconstruction technique (1.2 to 1.5)
- \( R_t \) at the center of the FOV: 5 mm for \( ^{18}\text{F} \)
Detection efficiency

\[ S = \frac{A}{4\pi r^2} \cdot \varepsilon^2 \text{[cps/MBq]} \]

- \( A \) = detector area seen by a point source to be imaged
- \( \varepsilon = 1 - \exp(-\mu x) \) detector’s efficiency
- \( \mu \) = linear attenuation coefficient of 511 keV photons in the detector
- \( x \) = thickness of the detector
- \( r \) = radius of the detector ring

- \( S = 0.2 \text{ – } 0.5 \% \) for 2D PET and 1-10\% for 3D PET
- \( (S = 0.01\text{-}0.03\% \text{ for SPECT}) \)

- Manufacturer provides volume sensitivity \( S_{\text{vol}} \) [cps/Bq/ml]
Coincidence event type in PET
Coincidence event type in PET

- **True coincidence**
  - Correct localization along the LOR
  - Useful for image reconstruction

- **Scatter coincidence (Compton)**
  - Mislocalization
  - Contrast reduction
  - Quantitative bias

- **Random coincidence**
  - Mislocalization
  - Important component → count rate saturation
  - Quantitative bias
Corrections for Quantitative Studies: All PET is (almost) Quantitative!

Raw Sinogram Data (Trues + Scatters + Randoms)

- Remove Randoms
- Normalize Detector Responses
- Correct for Deadtime
- Correct for Scatter
- Correct for Attenuation

It’s not exactly like this, and it’s not necessarily as linear as this!

T Sinogram Ready for Reconstruction
PVE correction

Recovery coefficient (RC)

$$RC = \frac{\text{Observed activity concentration}}{\text{True activity concentration}}$$

$$A_{\text{corrected}} = \frac{A_{\text{measured}}}{RC}$$

$^{18}\text{F (GE Discovery LS)}$
Specific Uptake Value (SUV) concept

\[
SUV = \frac{\text{Measured Activity Concentration \ [Bq/ml] \times 10^3}}{\frac{\text{Injected Activity}^* \ [Bq]}{\text{Patient Mass \ [kg]}}}
\]

\* Reported at the beginning of the exam

SUV = semi-quantitative index of the \(^{18}\text{FDG accumulation}\) makes possible the comparisons between exams

- SUV is used as an index to determine if a hotspot is significant
- Its use depends on:
  - Time between injection and acquisition, patient’s blood sugar level,
  - Patients’ weight, quantification quality (attenuation correction, …), partial volume effects, …
Recent developments
Acquisition geometry of the D-SPECT system with 9 detector blocks (A). Photograph of the camera (B), 1 detector column (C), and 1 CZT detector element (D). (SpectrumDynamics)

Acquisition geometry of the GE NM 530c multipinhole system (A). Photograph of the camera (B), multipinhole collimator (C)
The **CZT-SPECT-camera** (pixelated detector of a size of 2.46 mm/pixel and energy resolution of 6.3% compared to an energy resolution of 9.8% for the NaI-camera)

- Better energy resolution → reduce scatter contribution
- Higher spatial resolution ~ 3mm
- Higher count rate achievable
- Higher sensitivity
- → Shorter acq. time and/or low administered activity (patient dose reduction)

*Source: GE*
Semiconductor based (SiPM) PET

Detector modules developed to operate in high magnetic field (PET/MRI)

Source: GE
SiPM

Source: Philips

Conventional PM

Semiconductor SiPM

Improved signal localisation
Improved space resolution
Decreased signal Pile-up
Higher count rate achievable

Source: Philips
# Conventional vs. SiPM PET

## Table - Comparison of the Philips Ingenuity TF, GE Discovery 710, Biograph mCT Flow and the new Philips digital PET/CT Vereos.

<table>
<thead>
<tr>
<th>Model Product Name</th>
<th>Ingenuity TF</th>
<th>Discovery 710</th>
<th>Biograph mCT Flow</th>
<th>Vereos</th>
</tr>
</thead>
<tbody>
<tr>
<td>Patient port [cm]</td>
<td>70 OpenView</td>
<td>70</td>
<td>78</td>
<td>70</td>
</tr>
<tr>
<td>Patient scan range [cm]</td>
<td>190</td>
<td>200</td>
<td>195</td>
<td>190</td>
</tr>
<tr>
<td>Maximum patient weight [kg (lb)]</td>
<td>195 (430)</td>
<td>226 (500)</td>
<td>226 (500)</td>
<td>195 (430)</td>
</tr>
<tr>
<td>Acquisition modes</td>
<td>3D S&amp;S</td>
<td>3D S&amp;S</td>
<td>3D S&amp;S, continuous</td>
<td>3D S&amp;S</td>
</tr>
<tr>
<td>Number of image planes</td>
<td>45 or 90</td>
<td>47</td>
<td>109</td>
<td>72</td>
</tr>
<tr>
<td>Plane spacing [mm]</td>
<td>2 or 4</td>
<td>3.27</td>
<td>2</td>
<td>1, 2, or 4</td>
</tr>
<tr>
<td>Crystal size [mm]</td>
<td>4 x 4 x 22</td>
<td>4.2 x 6.3 x 25</td>
<td>4 x 4 x 20</td>
<td>4 x 4 x 22</td>
</tr>
<tr>
<td>Number of crystals</td>
<td>28,336</td>
<td>13,824</td>
<td>32,448</td>
<td>23,040</td>
</tr>
<tr>
<td>Number of PMTs</td>
<td>420</td>
<td>256</td>
<td>768</td>
<td>23,040 SiPMs</td>
</tr>
<tr>
<td>Physical axial FOV [cm]</td>
<td>18</td>
<td>15.7</td>
<td>21.8</td>
<td>16.3</td>
</tr>
<tr>
<td>Detector material</td>
<td>LYSO</td>
<td>LYSO</td>
<td>LSO</td>
<td>LYSO</td>
</tr>
<tr>
<td>System sensitivity 3D, [%]</td>
<td>0.74</td>
<td>0.75</td>
<td>0.85</td>
<td>&gt;1.0</td>
</tr>
<tr>
<td>Trans axial resolution @ 1 cm [mm]</td>
<td>4.7</td>
<td>4.9</td>
<td>4.4</td>
<td>4.0</td>
</tr>
<tr>
<td>Trans axial resolution @ 10 cm [mm]</td>
<td>5.2</td>
<td>5.5</td>
<td>4.9</td>
<td>4.5</td>
</tr>
<tr>
<td>Axial resolution @ 1 cm [mm]$^*$</td>
<td>4.7</td>
<td>5.6</td>
<td>4.5</td>
<td>4.0</td>
</tr>
<tr>
<td>Axial resolution @ 10 cm [mm]$^*$</td>
<td>5.2</td>
<td>6.3</td>
<td>5.9</td>
<td>4.5</td>
</tr>
<tr>
<td>Peak NECR [kcps]</td>
<td>120 @19 kBq/ml</td>
<td>130 @29.5 kBq/ml</td>
<td>175 @28 kBq/ml</td>
<td>400 @30 kBq/ml</td>
</tr>
<tr>
<td>Time-of-flight resolution [picoseconds]</td>
<td>591</td>
<td>544</td>
<td>540</td>
<td>307</td>
</tr>
<tr>
<td>Time-of-flight localization [cm]</td>
<td>8.9</td>
<td>8.2</td>
<td>8.1</td>
<td>4.6</td>
</tr>
<tr>
<td>Coincidence window [nanoseconds]</td>
<td>4.5</td>
<td>4.9</td>
<td>4.1</td>
<td>1.5</td>
</tr>
</tbody>
</table>

The sensitivity, NECR (noise equivalent count rate), coincidence window and TOF resolution are higher for the digital PET/CT digital PET/CT.

Abbreviations: FOV, field-of-view; PMT, photomultiplier tubes; NECR, noise equivalent count rate; kcps, kilocounts per second; kBq/ml, kiloBecquerel/milliliter; S&S, step and shoot.
