Imaging and detectors for medical physics

Lecture 5: Gamma cameras

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# Course layout

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<th>PM 15.30 – 17.00</th>
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<td>Lecture 2: Detectors for medical imaging</td>
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</table>
1. N Barrie Smith & A Webb  
   Introduction to Medical Imaging  
   Cambridge University Press

2. Edited by M A Flower  
   Webb’s Physics of Medical Imaging  
   CRC Press

3. A Del Guerra  
   Ionizing Radiation Detectors for Medical Imaging  
   World Scientific

4. W R Leo  
   Techniques for Nuclear and Particle Physics Experiments  
   Springer-Verlag
The gamma camera

Ref. 1 – Chapter 3.6

• Gamma camera = instrumental basis for:
  1. SPECT
  2. Planar scintigraphy

• → used in nuclear medicine imaging = emission imaging = radioactive source inside patient’s body

• In both SPECT and planar scintigraphy radiation source used = mainly $^{99}\text{Tc}m \rightarrow 140$ keV $\gamma$-rays → hence the name gamma camera

• Patient lies beneath gamma camera positioned close to organ of interest
Gamma camera operation

- Detects $\gamma$-rays and for each $\gamma$-ray records:
  1. x-y position
  2. Energy
  3. Time → can be linked to other information (for ex. ECG)

- Main features:
  1. Capable of withstanding $\gamma$-ray detection rates up to tens to thousands events per second
  2. Reject $\gamma$-rays scattered in the body → no useful spatial information
  3. As high sensitivity as possible → high quality images within clinically acceptable imaging time
Gamma camera components and operation

- **Gamma camera components:**
  1. Collimator
  2. Scintillator
  3. Array of PMTs
  4. ‘Anger’ position network
  5. Pulse height analyser (PHA)
  6. Digitizer = ADCs
  7. Computer to build up the image from many detection events

- **Operation:**
  1. $\gamma$-rays from patient travelling in certain directions selected
  2. $\gamma$-rays produce scintillation
  3. Scintillation detected by PMTs
Collimator

- $\gamma$-rays from radioactive source within the body emitted in all directions $\rightarrow$ high degree of collimation needed

- Operating principle: absorptive collimation
  Only $\gamma$-rays travelling in certain directions go through and reach the detector, the others are absorbed in the septa $\rightarrow$ most radiation is absorbed $\rightarrow$ inefficient

- Made usually from tungsten or lead to provide high absorption probability
Collimator design

- Array of holes drilled or cast in tungsten / lead or shaped from tungsten / lead foils
- Hexagonally-based ‘honeycomb’ geometry
- Septal thickness chosen to maximise attenuation (95%) and efficiency (sensitivity)
Collimator basic designs

- Six basic designs:
  1. Parallel-hole: most common design
  2. Converging
  3. Diverging
  4. Pinhole
  5. Slanthole
  6. Fan-beam
Parallel-hole collimator

- Holes are all parallel to each other
- Provides not magnified, not inverted images
- Comes in a variety of forms in which hole size, length and septal thickness are traded off to match energy of $\gamma$-ray

$I = \text{image object size}$
$O = \text{object size}$

Magnification $M = \frac{I}{O} = 1$
Parallel-hole collimator resolution

- Resolution of collimator $\delta r_{coll}$ is given by:

$$\delta r_{coll} \approx d \frac{l + z}{l}$$

- $d = \text{hole size}$
- $l = \text{septal length}$
- $z = \text{distance object – collimator}$
  = depth within the body

- Resolution:
  1. Independent of septal thickness $t$
  2. Gets worst with $z = \text{depth in body}$ and camera distance from body

- Courtesy Piero Posocco (Imperial College)
Parallel-hole collimator efficiency

- Geometric efficiency $G$ defined as fraction of $\gamma$-rays detected = transmitted by collimator divided by total number of $\gamma$-rays emitted

  $$G \approx \frac{d^4}{12l^2(d + t)^2}$$

- Assumption: attenuation in body ignored
- Efficiency independent of $z$

$d =$ hole size
$t =$ septa thickness
$l =$ septa thickness
$z =$ distance sourced - collimator
Parallel-hole collimator: resolution –vs– efficiency

• Bigger hole size → worst spatial resolution but higher efficiency → trade-off between the two:

\[ \delta r_{coll} \approx d \frac{l + z}{l} \quad G = \frac{d^4}{12l^2(d + t)^2} \]

• Example:

<table>
<thead>
<tr>
<th>Parameters</th>
<th>High resolution</th>
<th>High efficiency</th>
</tr>
</thead>
<tbody>
<tr>
<td>Septa thickness ( t ) (mm)</td>
<td>~0.2</td>
<td>~0.2</td>
</tr>
<tr>
<td>Septal length ( l ) (mm)</td>
<td>24</td>
<td>24</td>
</tr>
<tr>
<td>Hole size ( d ) (mm)</td>
<td>2</td>
<td>4</td>
</tr>
<tr>
<td>Spatial resolution ( \delta r_{coll} ) (mm) (^1)</td>
<td>10</td>
<td>21</td>
</tr>
<tr>
<td>Efficiency ( G ) (%)</td>
<td>( 4.8 \times 10^{-4} )</td>
<td>( 2.1 \times 10^{-3} )</td>
</tr>
</tbody>
</table>

\(^1\)For \( z = 100 \) mm
Use of parallel-hole collimators with radioisotopes

• Used with $^{99}Tc^m \ (E_\gamma = 140 \text{ keV})$ called low-energy collimators:
  1. Smaller holes $\rightarrow$ called low-energy high-resolution (LEHR) collimators
  2. Large holes $\rightarrow$ called low-energy all-purpose (LEAP) collimators

• Used with $^{11}Ga/^{111}I$ and $^{131}I$ called medium-, high-energy collimators:
  1. Septa of increasing thickness
Converging collimator

- Holes converge to a point in front of collimator = focused toward body
- Provides magnified images
- Focal point typically 40÷50 cm in front of collimator:
  1. For objects between collimator and convergence point → magnified non-inverted image
  2. Beyond convergence point → magnified inverted image
- Centre of curvature ideally in the middle of imaging FOV

Detector

Collimator

\[ M = \frac{I}{O} = \frac{f+l}{f-z} \]

\( I/O = \) image / object size
\( l = \) collimator thickness
\( z = \) distance object – collimator, \( f = \) focal distance

Courtesy Piero Posocco
(Imperial College)
Diverging collimator

- Converging collimator turned around → holes diverge from detector face = focused away from body
- Focal point typically 40÷50 cm behind collimator
- Provides de-magnified non-inverted images of size dependent on distance → distortions
- Larger FOV than parallel hole collimator
- Use: whole-body planar scintigraphy

De-magnification \( M = \frac{l}{O} = \frac{f-l}{f+z} \)
Pinhole collimator

- Single hole with interchangeable inserts of 3, 4, 6 mm aperture in lead, tungsten, gold, etc.
- Provides inverted images with significant (de)-magnification and high spatial resolution at small $z$ but also geometric distortion at image edges due to magnification changing with distance:
  1. For $z < f$ image magnified
  2. For $z > f$ image de-magnified
- Use: image very small organs such as thyroid and parathyroid

$De$-magnification $M = \frac{I}{O} = \frac{f}{z}$

$I/O = \text{image/ object size}$
$z = \text{distance object – collimator}$
$f = \text{focal distance}$

Courtesy Piero Posocco (Imperial College)
Slanthole collimator

- Parallel septa all tilted at same angle with respect to detection scintillator crystals
- Use: primarily in breast and cardiac imaging
Fan-beam collimator

• Two different geometries in the two directions:
  1. Head/foot direction = holes are parallel → parallel-hole collimator
  2. Radial direction = holes are converging → converging collimator

• Provides image magnification over reduced FOV
• Use: primarily for brain and heart studies
Scintillator

- Scintillation detector = large single $NaI(Tl)$ crystal of dimensions:
  1. Diameter = $40\div50$ cm
  2. Thickness = $6\div12.5$ cm chosen as compromise = the thicker the crystal:
     a. Broader LSF $\rightarrow$ lower spatial resolution
     b. Higher number of $\gamma$-rays detected $\rightarrow$ higher $SNR$

- $NaI(Tl)$ hygroscopic $\rightarrow$ needs hermetic sealing:
  1. Aluminium outside
  2. Glass toward PMT
### NaI(Tl) characteristics

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Value</th>
</tr>
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<tbody>
<tr>
<td>Linear attenuation coefficient at 140 keV</td>
<td>2.2 cm⁻¹</td>
</tr>
<tr>
<td>Decay time</td>
<td>230 ns</td>
</tr>
<tr>
<td>Peak emission wavelength</td>
<td>415 nm</td>
</tr>
<tr>
<td>Photon yield</td>
<td>38 $\gamma$/keV</td>
</tr>
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</table>

- Transparent to 415 nm light $\rightarrow$ little or no energy lost due to reabsorption
- Emission wavelength (415 nm) produced well-matched to optimal performance of conventional PMTs
- Photon yield (number of photons emitted) = directly proportional to $E_\gamma$
- Large crystals grown easily $\rightarrow$ cheap
PMT array

- Standard PMTs 2÷3 cm in diameter → array of 61, 75 or 91 PMTs used to cover the large crystal area
- Most efficient geometry: hexagonally-close pack → same distance from centre of one PMT to centre of each neighbouring PMT → important to reconstruct position of scintillation event
- Thin optical coupling layer
PMT calibration

- If PMT response across array not uniform → image artefacts
- Level of non-uniformity tolerated:
  a. Up to 10% in planar scans
  b. < 1% in SPECT
- PMT calibrated using sample of known radioactivity → calibration coefficient calculated and applied to the data
  - New calibration systems use as calibration source one LED per PMT → continuous calibration also during scans
‘Anger’ position network

- Scintillation event in $NaI(Tl)$ $\rightarrow$ signal in PMTs:
  - PMT closest to event $\rightarrow$ largest output signal
  - Adjacent PMTs $\rightarrow$ smaller output signals = light detected $\sim$ inversely proportional to distance event – PMT

- Position of scintillation event calculated by comparing magnitudes of PMTs output signals = centroid or centre of mass of light distribution in PMTs

- More complex positioning algorithms that account for local spatial distortions and apply corrections for PMT non-uniform performance can also be used
‘Anger’ position network something extra

In simplest (linear) case the position is given by:

\[ D_1 = \frac{(S_2 \times D)}{(S_1 + S_2)} \]
\[ D_2 = \frac{(S_1 \times D)}{(S_1 + S_2)} \]

\( D_1, D_2 = \) distances scintillation event – centre of PMT
\( D = D_1 + D_2 \)
\( S_1, S_2 = \) PMT output signals
Pulse height analyser

• The pulse height analyser determines the recorded events where:
  
  1. The $\gamma$-ray has not been scattered in the patient = good event → to be retained
  2. The $\gamma$-ray has been Compton scattered in the patient → lost spatial information = bad event → to be rejected

• The pulse height analyser allows spectroscopic imaging = selects for display only events falling within a particular energy range
Pulse height analyser operating principle

• Amplitude of PMT output signal proportional to $E_\gamma$:
  1. Amplitude of PMT output signal proportional to number of optical photons arriving from scintillator
  2. Number of optical photons produced in scintillator proportional to $E_\gamma$

• Discriminating on the basis of amplitude of PMT output signal = discriminating on the basis of $E_\gamma$

• The pulse height analyser takes summed electrical signals from the PMT array and applies thresholds
Pulse height analyser operation

• Two steps:
  1. The PMT output signals are sent to the multiple-channel analyser → pulse-height histogram
  2. Two voltage thresholds, upper and lower, are applied to the pulse-height histogram to select the $\gamma$-rays to be accepted
Pulse height analyser
multiple-channel analyser

• Multiple-channel analyser (MCA):
  1. Receives in input PMT output signals
  2. ADC digitizes the signals
  3. Produces the pulse-height histogram = plot of number of events from the PMTs as function of output amplitude

• Channel = specific energy range → number of channels can be > 1000 → complete energy spectra produced

Courtesy Piero Posocco
(Imperial College)
Pulse height analyser thresholds

• Variation in $E_\gamma$ detected for various reasons:
  1. Non-uniformity in PMT and scintillator response $\rightarrow$ range of outputs for mono-energetic $\gamma$-rays
  2. Compton scattering of $\gamma$-rays $\rightarrow$ variable lower $E_\gamma$:
     a. Scattering in patient at small angles still has useful spatial information $\rightarrow$ event needs to be accepted
     b. Scattering in scintillator retains useful spatial information $\rightarrow$ event needs to be accepted

• One voltage level at photopeak = full $E_\gamma$ to select accepted event not enough $\rightarrow$ energy range = detection window needed $\rightarrow$ two thresholds
Detection window in context

- Isomeric transition of $^{99}Tc^m \rightarrow 140$ keV $\gamma$-ray
- Peaks in pulse-height diagram:
  - $A =$ total absorption of 140 keV $\gamma$-ray by scintillator
  - $B =$ iodine escape peak = $\gamma$-ray excites iodine $\rightarrow K$-shell X-ray emitted $\rightarrow$ escapes scintillator $\rightarrow$ detected $E_\gamma \ 28\div33$ keV lower
  - $C =$ photoelectric absorption in lead collimator $\rightarrow K$-shell X-rays at $\sim 78$ keV $\rightarrow$ absorbed in scintillator
- Detection window set to accept A and B but reject C

Courtesy Piero Posocco (Imperial College)
Types of events

- **A** = good event
- **B** = scatter in detector
  - $\gamma$-ray scatters in detector $\Rightarrow$ position information distorted
  - Full energy deposited $\Rightarrow$ accepted
- **C** = scatter in patient
  - $\gamma$-ray scatters in patient and may retain sufficient energy to fall in energy window
  - unwanted event
- **D** = septal penetration
  - $\gamma$-ray penetrates through wall of collimator and reaches detectors
  - unwanted event

Patient with $\gamma$-emitting radionuclide concentrated in liver

Courtesy Piero Posocco (Imperial College)
Gamma camera energy resolution

- Energy resolution defined as FWHM of photopeak → the narrower FWHM the better is the gamma camera at discriminating unscattered and scattered $\gamma$-rays
- Typical energy resolution values:
  1. Without patient: $14$ keV $= 10\%$ → measured during calibration
  2. With patient: $28$ keV $= 20\%$ → set at this value to increase the acceptance window
     Ex.: $20\%$ window around photopeak $= 20\%$ around $140$ keV → $\gamma$-rays in energy range $127\div153$ keV accepted
Gamma camera detection efficiency

- Gamma camera detection efficiency = fraction of emitted $\gamma$-rays that produce counts in the image
- The detection efficiency $E_{\text{camera}}$ made by three terms:

$$E_{\text{camera}} = G \cdot \varepsilon \cdot f$$

$G = \text{collimator contribution}$
$\varepsilon = \text{scintillator contribution}$
$f = \text{read-out contribution}$
Terms contributing to the detection efficiency

1. Collimator → geometric collimator efficiency

2. Scintillator → ratio of $\gamma$-rays recorded by the scintillator $N_{record}$ to the $\gamma$-rays hitting it $N_{hit}$:

$$\varepsilon = \frac{N_{record}}{N_{hit}} = 1 - \exp(-\mu \cdot d)$$

$\mu = \text{scintillator linear attenuation coefficient at } E_\gamma$
$\text{d = scintillator thickness}$

3. Read-out → fraction of photons accepted by energy discriminating unit
Dead-time

- If injected dose very large\(^1\) number of \(\gamma\)-rays arriving on scintillator can exceed system recording capabilities due to electronics finite recovery and reset times → events get lost during the time the system cannot respond = cannot record the events → dead-time

- Two types of behaviour:
  1. Paralysable
  2. Non-paralysable

- Dead-time determined by non-paralysable behaviour

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\(^1\)This is particularly true at start of scan
**Paralysable / non-paralysable behaviour**

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<thead>
<tr>
<th>Paralysable behaviour</th>
<th>Non-paralysable behaviour</th>
</tr>
</thead>
<tbody>
<tr>
<td>Component cannot respond to new event until after a certain time from previous event</td>
<td>Component cannot respond for a set time irrespective of level of radioactivity</td>
</tr>
</tbody>
</table>

Example:
Each time a $\gamma$-ray the scintillator produces excited state that decays after 230 ns. If another $\gamma$-ray strike the scintillator on the same spot before the excite state decay, it will take another 230 ns for the state to decay and only one set of photons will be produced. The higher the rate of $\gamma$-rays striking the detector, the longer the dead-time.

Example:
The PHA takes a certain time to process the input signal from an event. During this time further events are not recorded independently from how many they are.
Determination of dead-time

• The dead-time $\tau$ is given by:
  $$\tau = \frac{N - n}{nN}$$
  
  $N = \text{true count rate} = \text{number of scintillations / s}$
  $n = \text{measure count rate}$

• Maximum measurable count rate $= \frac{1}{\tau}$

• Dead-time corrections based on calibration using phantoms of known radioactivity
Gamma camera spatial resolution

- Total gamma camera spatial resolution $\delta r_{tot}$ given by sum of three terms:
  \[
  \delta r_{tot} = \sqrt{\delta r_{coll}^2 + \delta r_{int}^2 + \delta r_{Compton}^2}
  \]
  - $\delta r_{coll}$ = collimator spatial resolution
  - $\delta r_{int}$ = gamma camera intrinsic resolution
  - $\delta r_{Compton}$ = Compton scattering contribution
- Typical values for overall spatial resolution are:
  1. $1\div2$ cm deep in the body
  2. $5\div8$ cm close to surface of body → close to collimator
Terms contributing to spatial resolution

1. Collimator spatial resolution:
   a. Determined by:
      i. Collimator geometry
      ii. Depth of source = organ where radiotracer is accumulated
   b. Typical values 5÷8 mm
   c. Typically dominant term to total spatial resolution

2. Gamma camera intrinsic resolution
   a. Due to:
      i. Uncertainty in exact localisation of point where scintillation occurs = ability of PMTs to localise events
      ii. Intrinsic resolution of positioning system
   b. Typical values 3÷5 mm

2. Compton scattering contribution
   a. Due to $\gamma$-rays undergoing Compton scattering in the body
Signal-to-noise ratio

- Radioactive decay = statistical process → number of disintegrations per unit time fluctuates
- Poisson distribution → \( SNR \) depends on number of detected \( \gamma \)-rays \( N \):
  \[ SNR \propto \sqrt{N} \]

- Factors affecting \( SNR \):
  1. Administered radioactivity
  2. Radiotracer uptake
  3. Depth of organ in body = the deeper the organ the more attenuation → lower \( SNR \)
  4. Total scanning time
  5. Intrinsic resolution of the gamma camera → only collimator properties can be changed
  6. Post-acquisition filtering → increases \( SNR \) but adds blur
Contrast and contrast-to-noise ratio

• Theoretically no noise $\rightarrow$ contrast and $CNR$ very high
  No background from tissues of no interest = where radiotracer does not accumulated
• In reality Compton scattered $\gamma$-rays $\rightarrow$ random background signal higher close to region with uptake of radiotracer