Joint CI-JAI advanced accelerator lecture series

Imaging and detectors for medical physics

Lecture 5: Gamma cameras

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Course layout

Day	AM 09.30 – 11.00	PM 15.30 – 17.00
Week 1		
6th June	Lecture 1: Introduction to medical imaging	Lecture 2: Detectors for medical imaging
7th June	Lecture 3: X-ray imaging	
8th June		Tutorial
Week 2		
13th June	Lecture 4: Radionuclides	
14th June	Lecture 5: Gamma cameras	Lecture 6: SPECT
16th June	Lecture 7: PET	
Week 3		
22nd June	Tutorial	

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Books

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

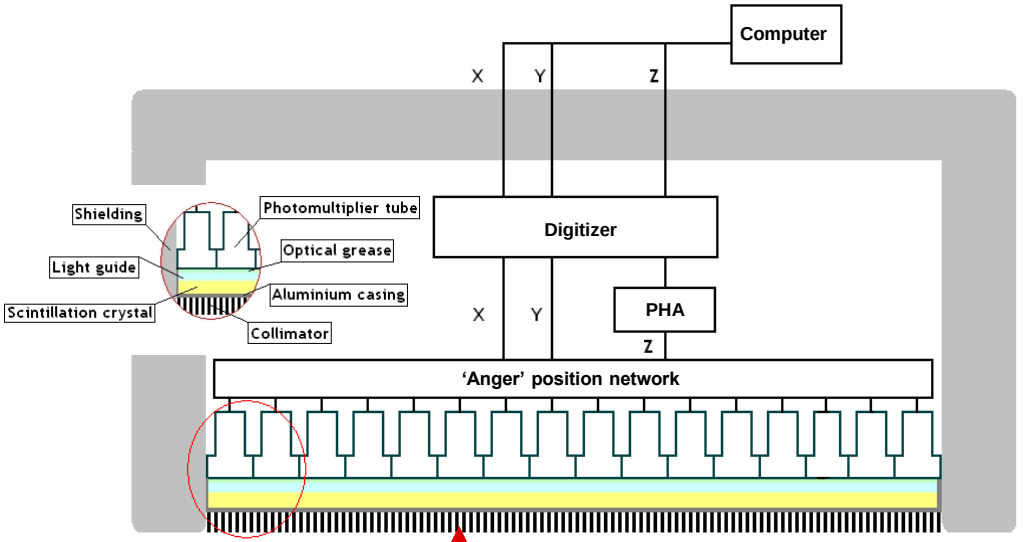
Ref. 1 – Chapter 3.6

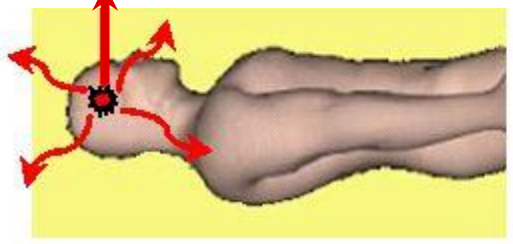
- Gamma camera= instrumental basis for:
  1. SPECT
  2. Planar scintigraphy
- $\rightarrow$ used in nuclear medicine imaging = emission imaging = radioactive source inside patient's body
- In both SPECT and planar scintigraphy radiation source used = mainly $^{99}Tc^{m} \rightarrow 140$ keV γ-rays $\rightarrow$ hence the name gamma camera
- Patient lies beneath gamma camera positioned close to organ of interest

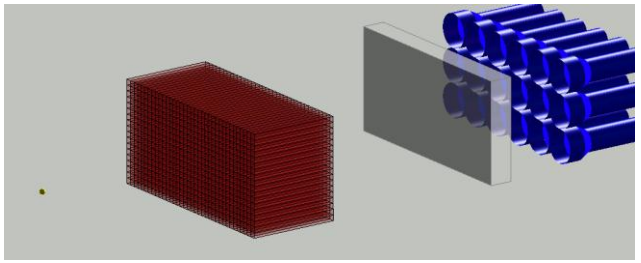
Gamma camera operation

- Detects γ-rays and for each γ-ray records:
  1. x-y position
  2. Energy
  3. Time $\rightarrow$ can be linked to other information (for ex. ECG)
- Main features:
  1. Capable of withstanding γ-ray detection rates up to tens to thousands events per second
  2. Reject γ-rays scattered in the body $\rightarrow$ no useful spatial information
  3. As high sensitivity as possible $\rightarrow$ high quality images within clinically acceptable imaging time

Gamma camera components and operation







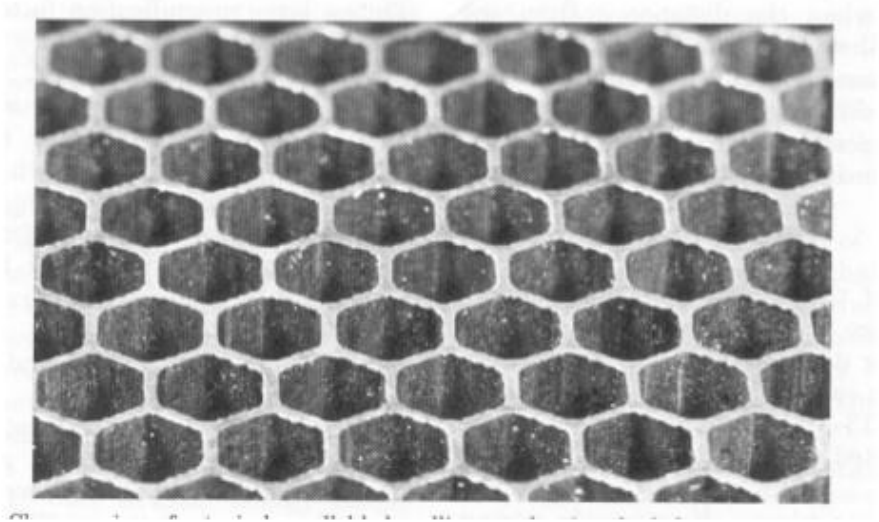
- Gamma camera components:
  1. Collimator
  2. Scintillator
  3. Array of PMTs
  4. 'Anger' position network
  5. Pulse height analyser (PHA)
  6. Digitizer = ADCs
  7. Computer to build up the image from many detection events
- Operation:
  1. γ-rays from patient travelling in certain directions selected
  2. γ-rays produce scintillation
  3. Scintillation detected by PMTs

Collimator

- γ-rays from radioactive source within the body emitted in all directions → high degree of collimation needed
- Operating principle: absorptive collimation

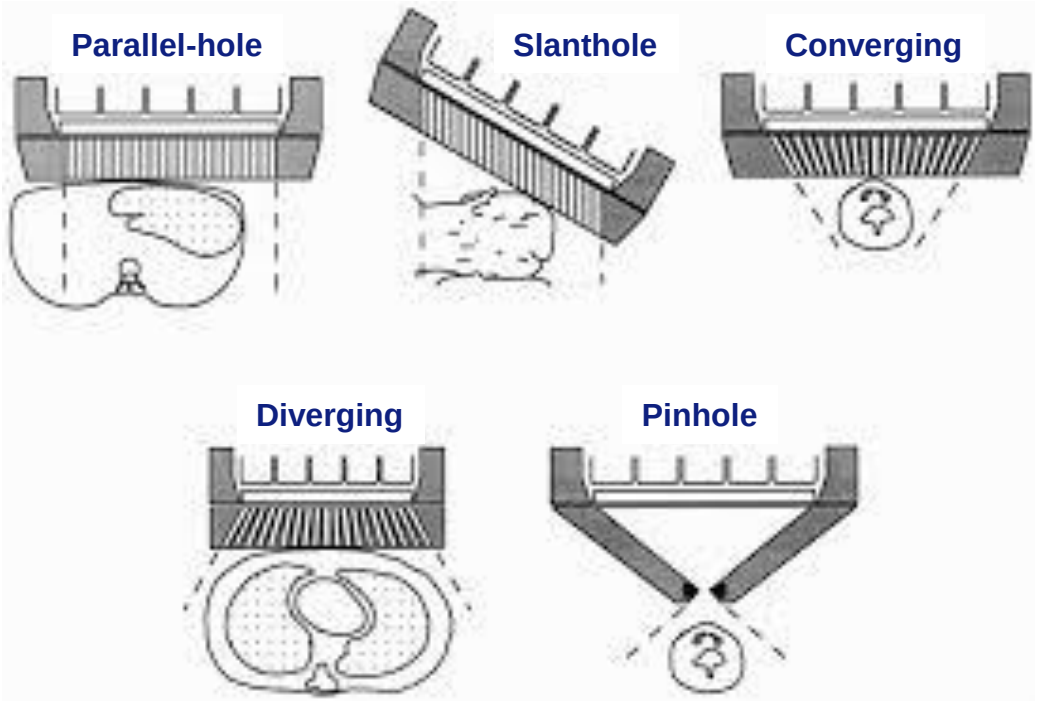
  Only γ-rays travelling in certain directions go through and reach the detector, the others are absorbed in the septa → most radiation is absorbed → inefficient
- Made usually from tungsten or lead to provide high absorption probability

Collimator design



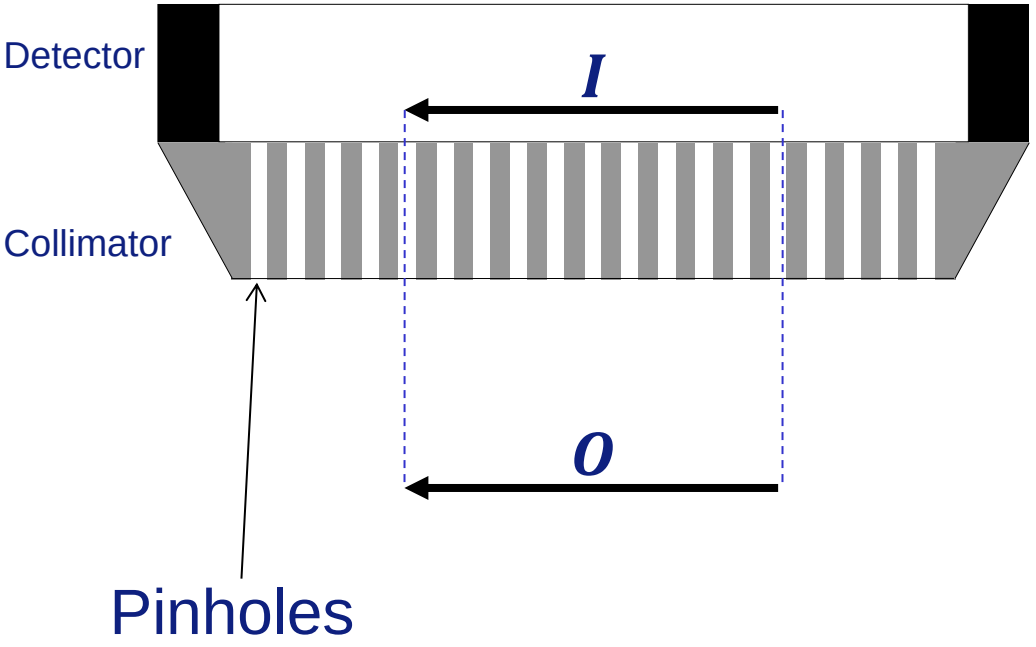
- Array of holes drilled or cast in tungsten / lead or shaped from tungsten / lead foils
- Hexagonally-based ‘honeycomb’ geometry
- Septal thickness chosen to maximise attenuation (95%) and efficiency (sensitivity)

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



Courtesy Piero Posocco (Imperial College)

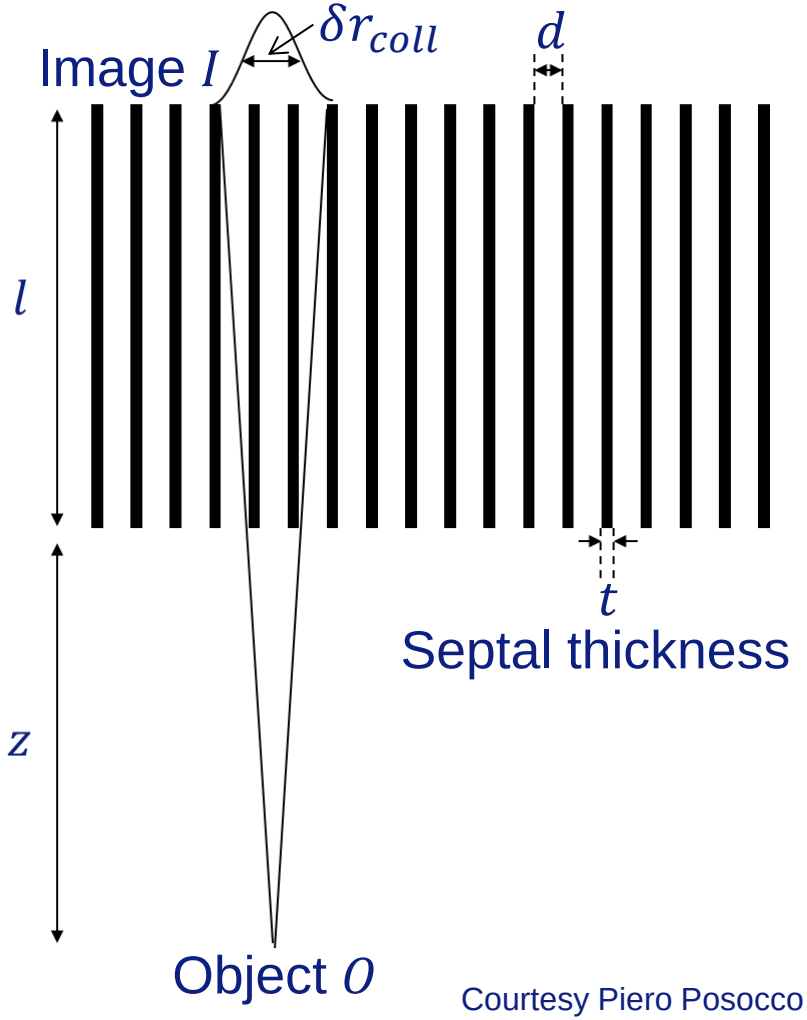
- Holes are all parallel to each other
- Provides not magnified, not inverted images
- Comes in a variety of forms in which hole size, length and septal thickness are traded off to match energy of γ-ray

I = image object size

O = object size

$$\text{Magnification } M = \frac{I}{O} = 1$$

Parallel-hole collimator resolution



Courtesy Piero Posocco (Imperial College)

- Resolution of collimator δr_{coll} is given by:

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

d = hole size

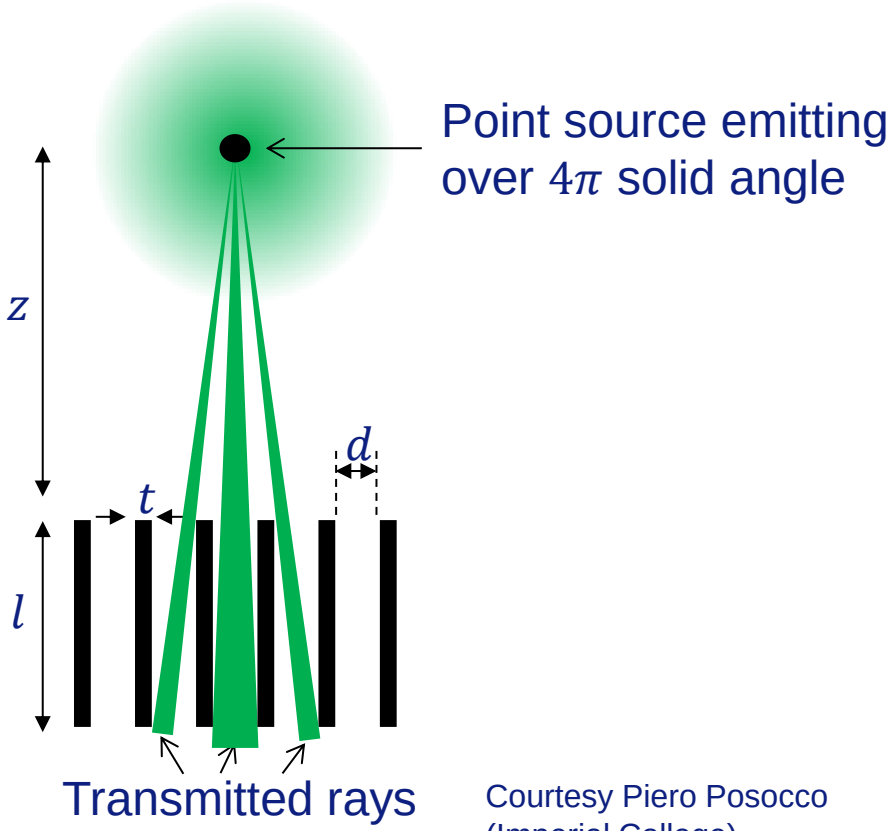
l =septal length

z =distance object – collimator

= depth within the body

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

Parallel-hole collimator efficiency



Courtesy Piero Posocco (Imperial College)

d = hole size
t = septa thickness
l = septa thickness
z = distance sourced - collimator

- Geometric efficiency G defined as fraction of γ-rays detected = transmitted by collimator divided by total number of γ-rays emitted
- Given by:

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

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

Parallel-hole collimator: resolution –vs– efficiency

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

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

- Example:

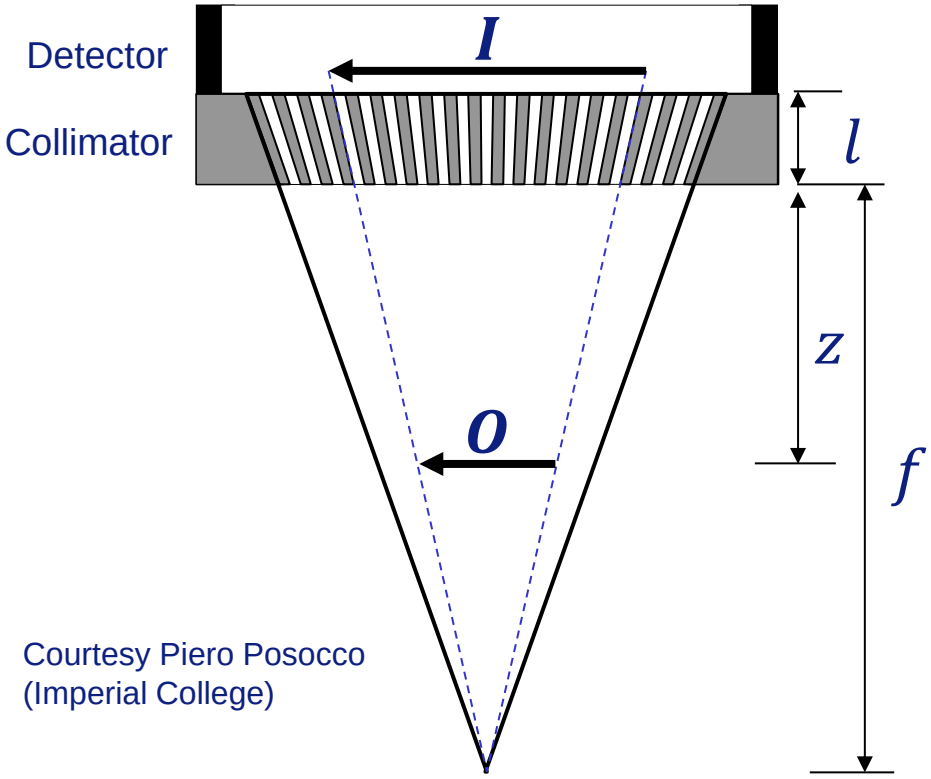
Parameters	High resolution	High efficiency
Septa thickness t (mm)	~ 0.2	~ 0.2
Septal length l (mm)	24	24
Hole size d (mm)	2	4
Spatial resolution δr_{coll} (mm) [1]	10	21
Efficiency G (%)	4.8×10^{-4}	2.1×10^{-3}

[1]For $z = 100$ mm

Use of parallel-hole collimators with radioisotopes

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



Courtesy Piero Posocco
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I/O = image / object size

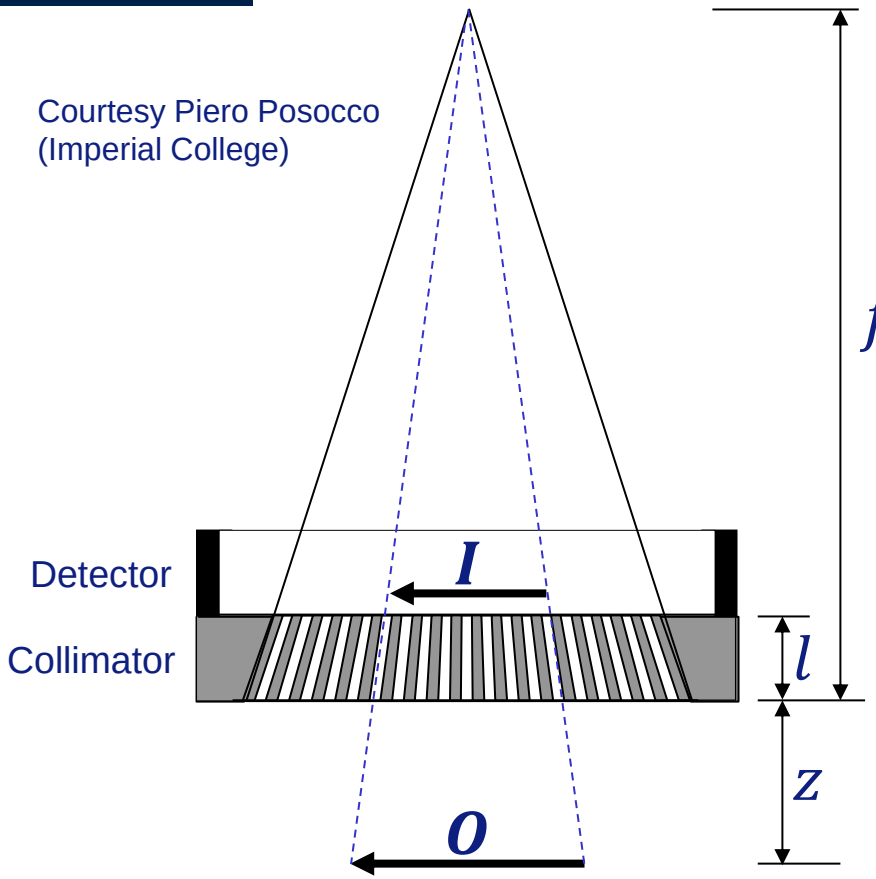
l = collimator thickness

z = distance object – collimator, f = focal distance

Magnification $M = \frac{I}{O} = \frac{f+l}{f-z}$

- Holes converge to a point in front of collimator = focused toward body
- Provides magnified images
- Focal point typically $40 \div 50$ 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

Diverging collimator

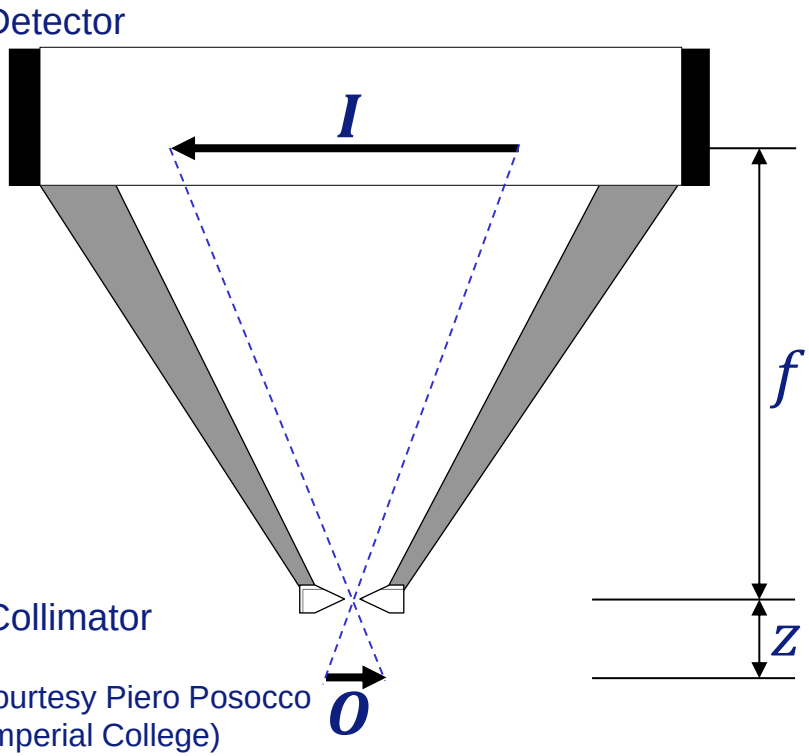


I, O, l, z, f = as in previous slide

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

- Converging collimator turned around → holes diverge from detector face = focused away from body
- Focal point typically $40 \div 50$ 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

Pinhole collimator



Courtesy Piero Posocco
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I/O = image / object size

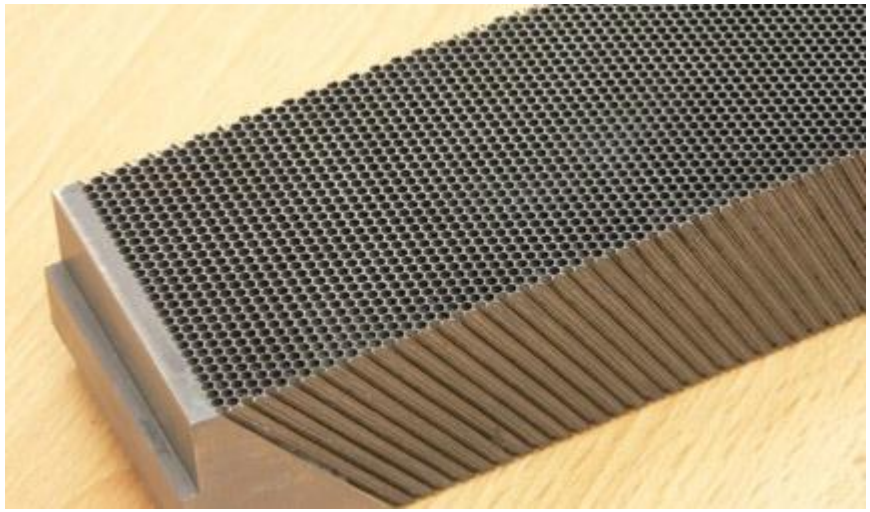
z = distance object – collimator,

f = focal distance

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

- Single hole with interchangeable inserts of $3,4,6$ mm aperture in led, 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

Slanthole collimator

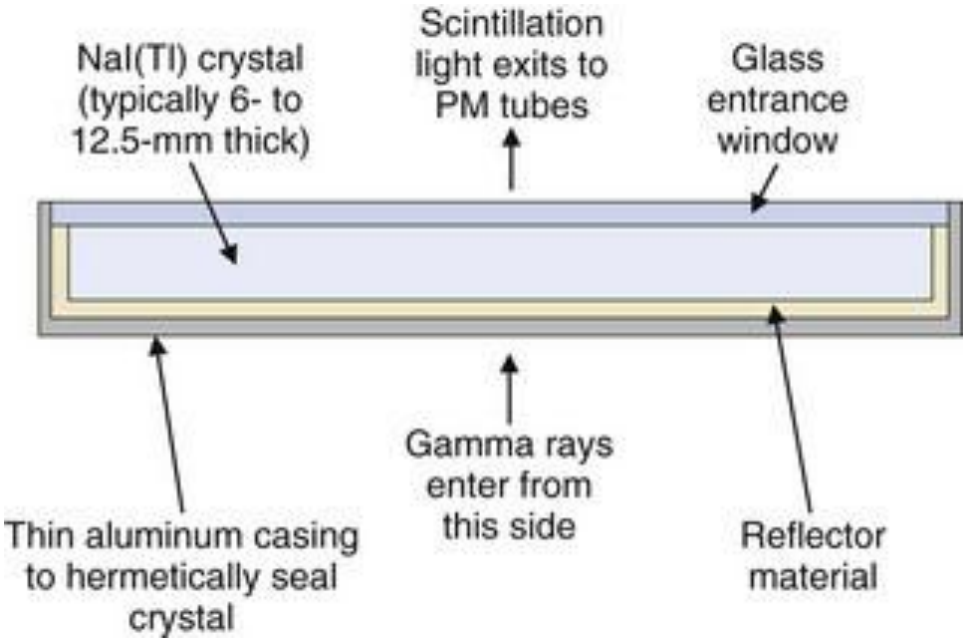


- Parallel septa all tilted at same angle with respect to detection scintillator crystals
- Use: primarily in breast and cardiac imaging

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 = $40 \div 50$ cm
  2. Thickness = $6 \div 12.5$ cm chosen as compromise = the thicker the crystal:
     a. Broader LSF → lower spatial resolution
     b. Higher number of γ-rays detected → higher SNR
- $NaI(Tl)$hygroscopic → needs hermetic sealing:
  1. Aluminium outside
  2. Glass toward PMT

$NaI(Tl)$ characteristics

Parameter	Value
Linear attenuation coefficient at 140 keV	$2.2\ \mathrm{cm}^{-1}$
Decay time	230 ns
Peak emission wavelength	415 nm
Photon yield	$38\ \gamma$/keV

- Transparent to 415 nm light → 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_γ
- Large crystals grown easily → cheap

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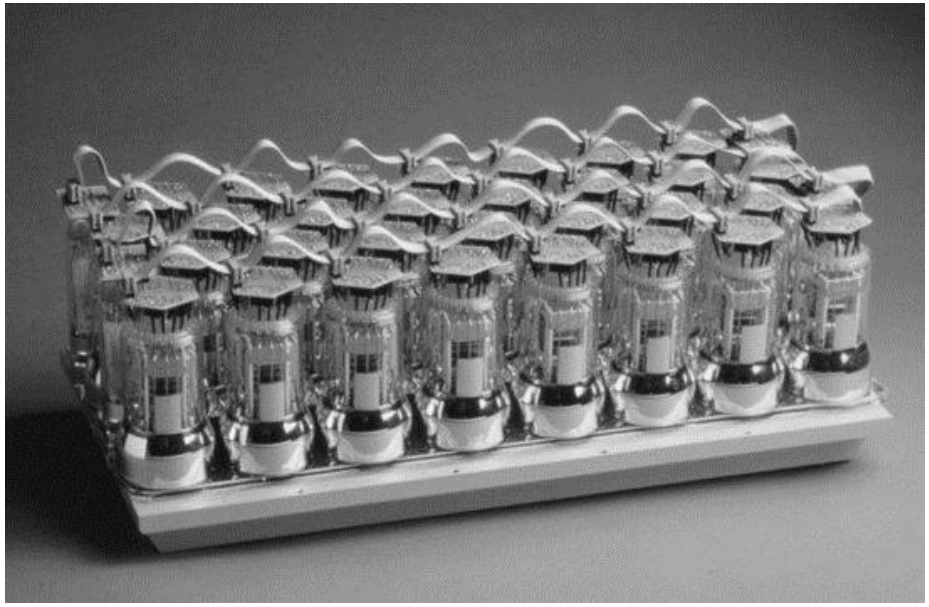
PMT array



- Standard PMTs $2 \div 3$ 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

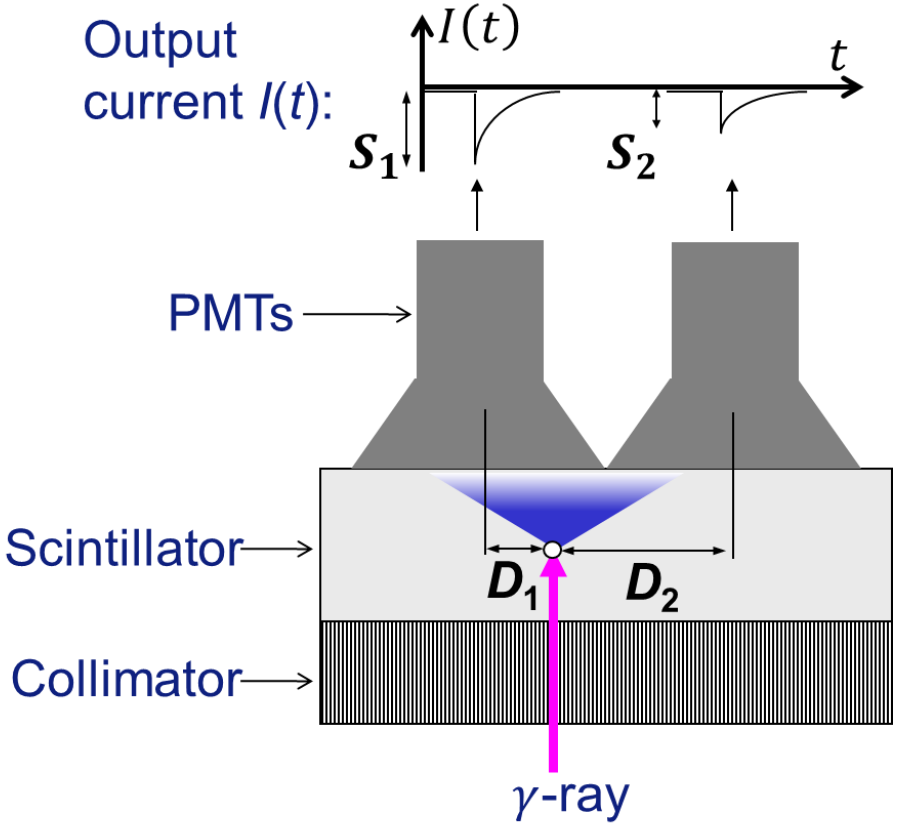
PMT calibration

- 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



Courtesy Piero Posocco (Imperial College)

- 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 γ-ray has not been scattered in the patient = good event $\rightarrow$ to be retained
  2. The γ-ray has been Compton scattered in the patient $\rightarrow$ lost spatial information = bad event $\rightarrow$ 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

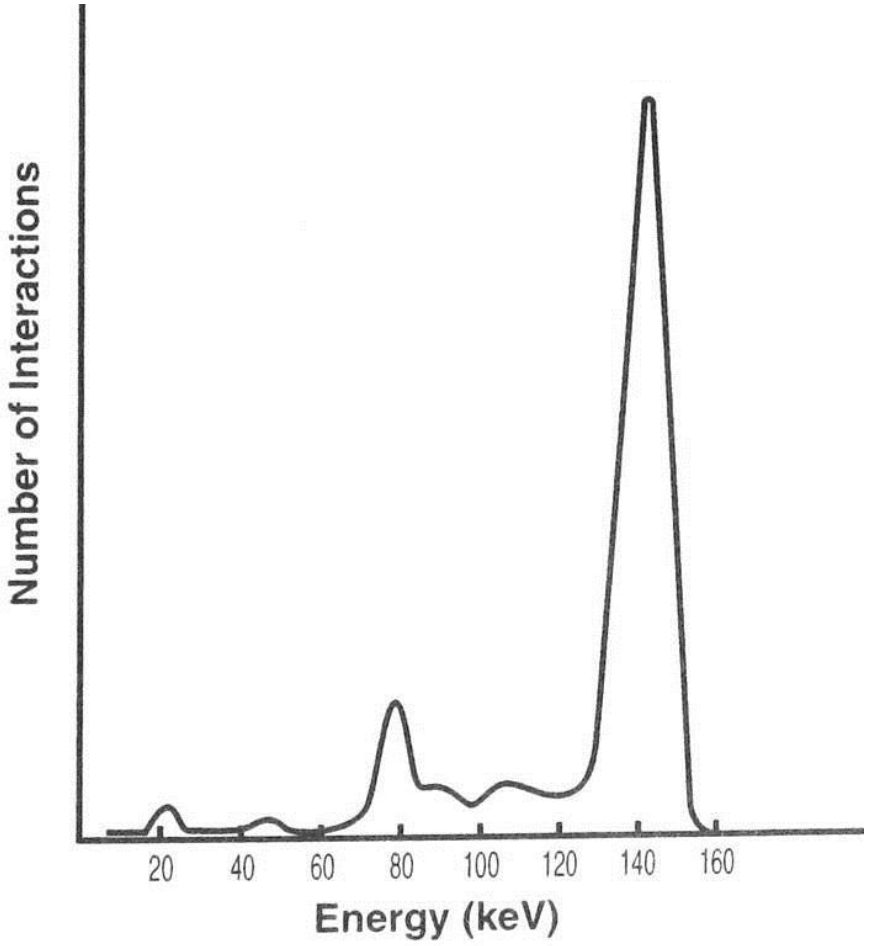
- Amplitude of PMT output signal proportional to E_γ:
  1. Amplitude of PMT output signal proportional to number of optical photons arriving from scintillator
  2. Number of optical photons produced in scintillator proportional to E_γ
- Discriminating on the basis of amplitude of PMT output signal = discriminating on the basis of E_γ
- 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 γ-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



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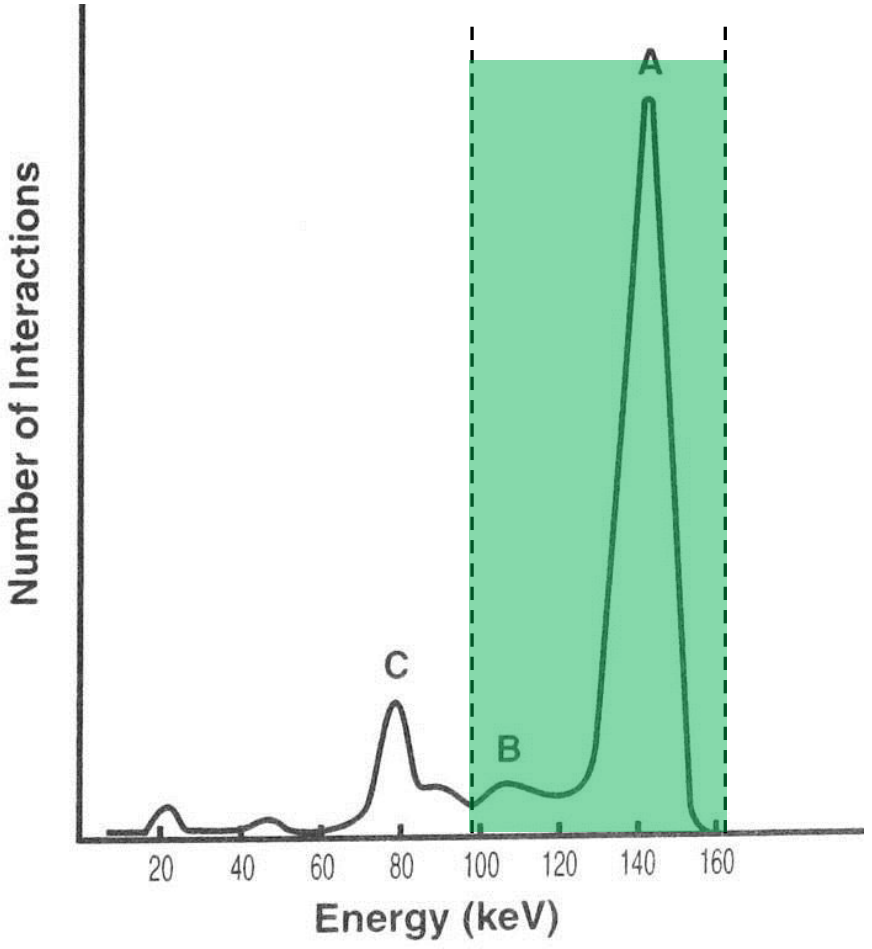
Pulse height analyser thresholds

- Variation in E_γ detected for various reasons:
  1. Non-uniformity in PMT and scintillator response → range of outputs for mono-energetic γ-rays
  2. Compton scattering of γ-rays → variable lower E_γ:
     a. Scattering in patient at small angles still has useful spatial information → event needs to be accepted
     b. Scattering in scintillator retains useful spatial information → event needs to be accepted
- One voltage level at photopeak = full E_γ to select accepted event not enough → energy range = detection window needed → two thresholds

Detection window in context

- Isomeric transition of $^{99}Tc^{m} \rightarrow$ 140 keV γ-ray
- Peaks in pulse-height diagram:
  - A = total absorption of 140 keV γ-ray by scintillator
  - B = Iodine escape peak = γ-ray excites iodine → K-shell X-ray emitted → escapes scintillator → detected E_γ 28÷33 keV lower
  - C = photoelectric absorption in lead collimator → K-shell X-rays at ~78 keV → absorbed in scintillator
- Detection window set to accept A and B but reject C



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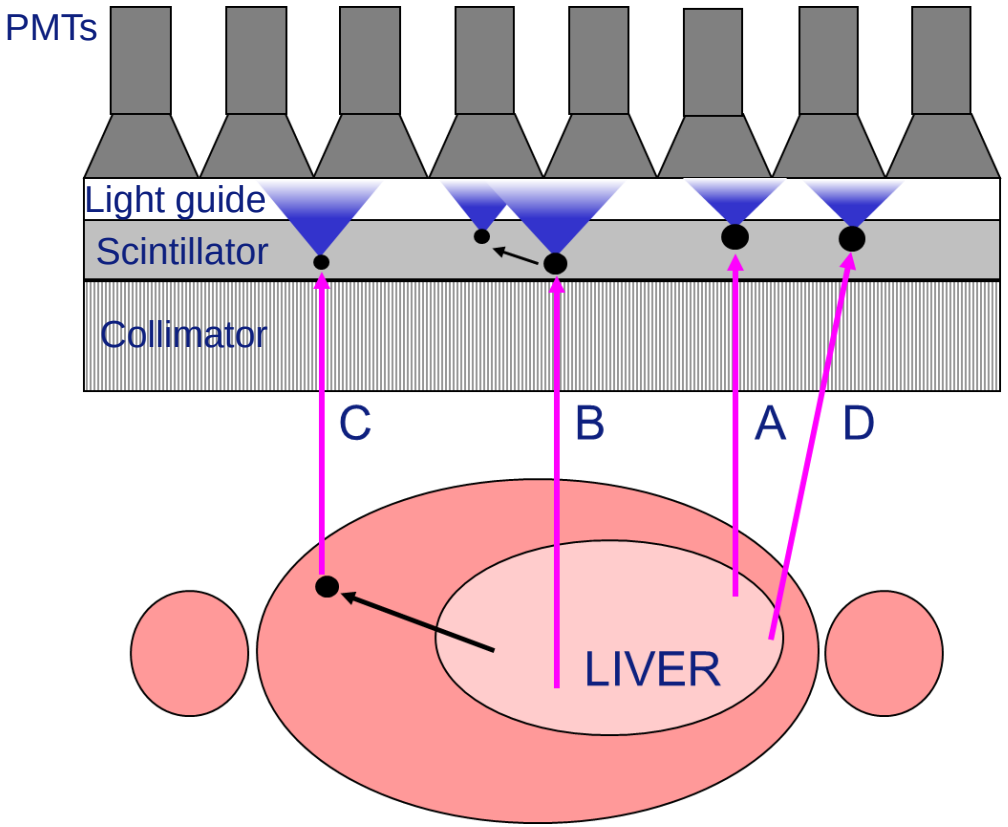
Types of events



Patient with γ-emitting radionuclide concentrated in liver

Courtesy Piero Posocco (Imperial College)

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

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 γ-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 → γ-rays in energy range $127 \div 153$ keV accepted

Gamma camera detection efficiency

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

$$\mathrm{E}_{camera} = G \cdot \varepsilon \cdot f$$

$G =$ collimator contribution

$\varepsilon =$ scintillator contribution

$f =$ read-out contribution

Terms contributing to the detection efficiency

1. Collimator → geometric collimator efficiency
2. Scintillator → ratio of γ-rays recorded by the scintillator N_{record} to the γ-rays hitting it N_{hit}:

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

   μ = scintillator linear attenuation coefficient at E_γ

   d = scintillator thickness

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

Dead-time

- If injected dose very large[1] number of γ-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

[1]This is particularly true at start of scan

Paralysable / non-paralysable behaviour

Paralysable behaviour	
Component cannot respond to new event until after a certain time from previous event	Example: Each time a γ-ray the scintillator produces excited state that decays after 230 ns. If another γ-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 γ-rays striking the detector, the longer the dead-time

Non-paralysable behaviour	
Component cannot respond for a set time irrespective of level of radioactivity	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 τ is given by:

$$\tau = \frac{N - n}{nN}$$

N = true count rate = number of scintillations / s

n = 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 δr_{tot} given by sum of three terms:

$$\delta r_{tot} = \sqrt{\delta r_{coll}^2 + \delta r_{int}^2 + \delta r_{Compton}^2}$$

δr_{coll} = collimator spatial resolution

δr_{int} = gamma camera intrinsic resolution

$\delta r_{Compton}$ = Compton scattering contribution

- Typical values for overall spatial resolution are:
  1. $1 \div 2$ cm deep in the body
  2. $5 \div 8$ 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 $5 \div 8$ 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 $3 \div 5$ mm
2. Compton scattering contribution
   a. Due to γ-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 γ-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 → contrast and CNR very high

  No background from tissues of no interest = where radiotracer does not accumulated
- In reality Compton scattered γ-rays → random background signal higher close to region with uptake of radiotracer