Lecture 11 – Gamma camera

This lecture will cover: *(CH3.6-3.7)*

- Instrumentations
  - Collimators
  - Detector scintillation crystal
  - Photomultipliers tubes
  - Anger position network
  - Pulse height analyzer
  - Instrument dead time
- Image characteristics
  - Influence of instrumentation
  - SNR
  - Spatial resolution
  - Contrast and contrast-to-noise

(*Supplementary reading: The Essential Physics of Medical Imaging CH17-18*)

Gamma Camera

- Components of gamma camera:
  - Collimator
  - Detector scintillation crystal and coupled photomultipliers tubes
  - Anger position network and pulse height analyzer
- Detecting γ-rays at the rate of up to tens of thousand per second
- Rejecting scattered γ-rays
- High sensitivity of detecting γ-rays for
  - High quality images
  - Clinically acceptable imaging time

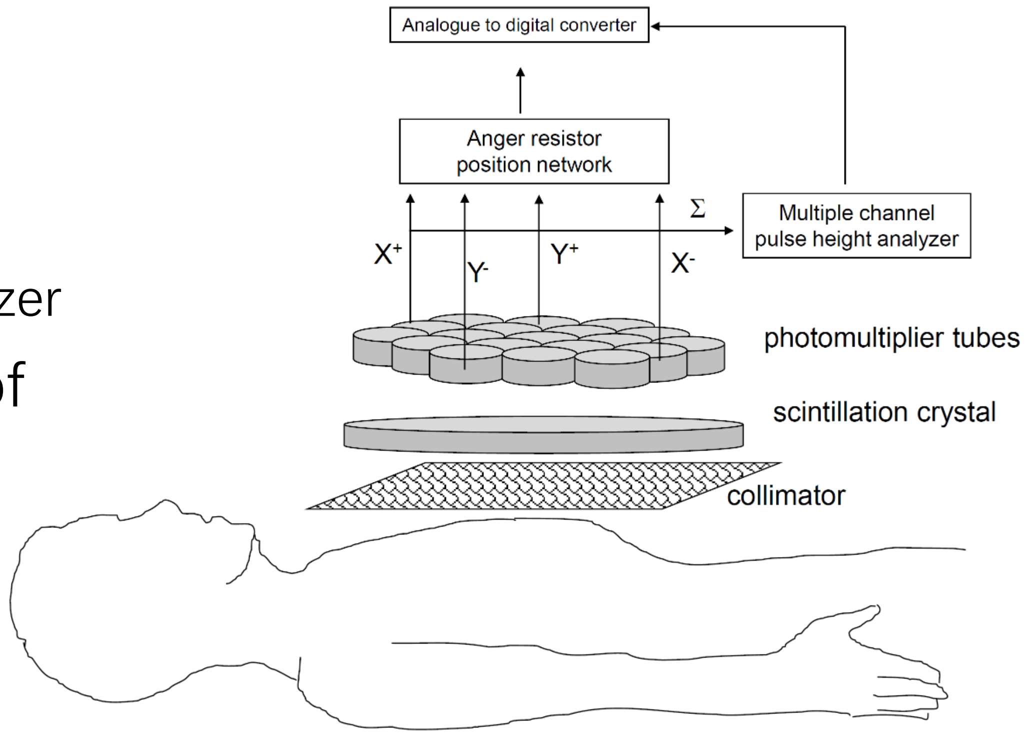


Fig. A gamma camera used for both planar scintigraphy (one camera) and SPECT (two or three rotating cameras).

Collimator

- Similar to the anti-scatter grid in X-ray imaging – thin strips of lead
- Higher degree of collimation required, and higher attenuation of γ-rays
- Collimator designs
  - Parallel hole collimator
  - Slant hole collimator
  - Converging collimator
  - Diverging collimator
  - Fan-beam collimator
  - Pinhole collimator

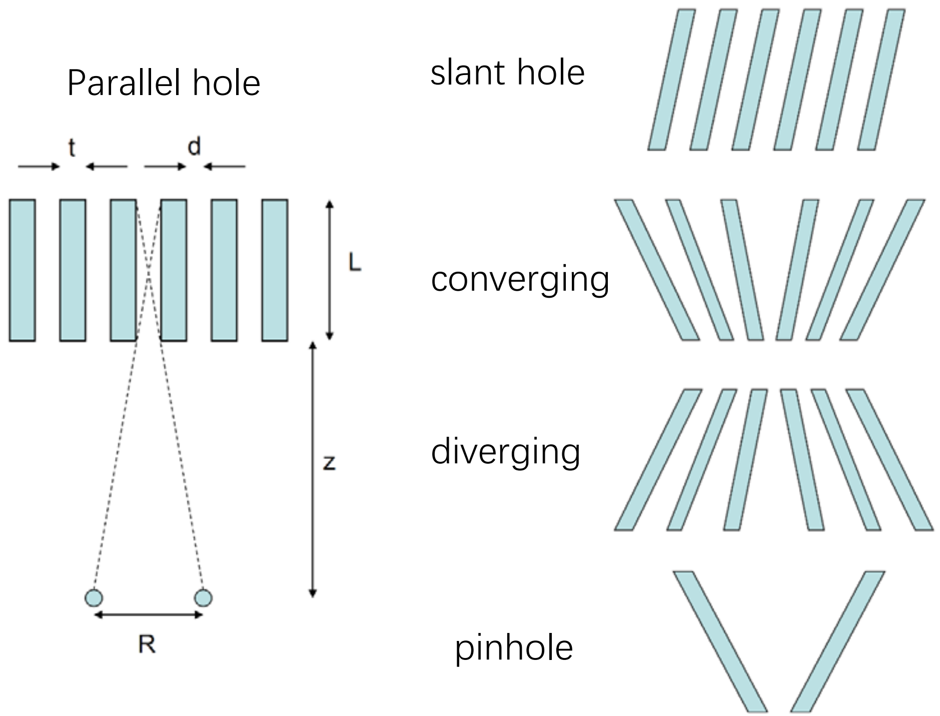


Fig. (left) A parallel hole collimator, with parameters necessary to calculate the spatial resolution, R. The patient would be positioned at the bottom of the figure, with the gamma camera at the top. (right) Four common types of collimator: from top-to-bottom, slanting hole, converging, diverging and pinhole.

Parallel hole collimator

- Hexagonally-based “honeycomb” geometry
- The spatial resolution of collimator: $R = \frac{d(L_{eff}+z)}{L_{eff}}$
  - The effective septal length: $L_{eff} = L - \frac{2}{\mu_{septa}}$
  - Better resolution for regions closer to the surface than those deeper in the body
- The efficiency of collimator:

$$g \approx K^2 \frac{d^2}{L_{eff}{}^2} \frac{d^2}{(d+t)^2}$$

Where K depends on the particular hole-geometry, and 0.26 for hexagonal holes

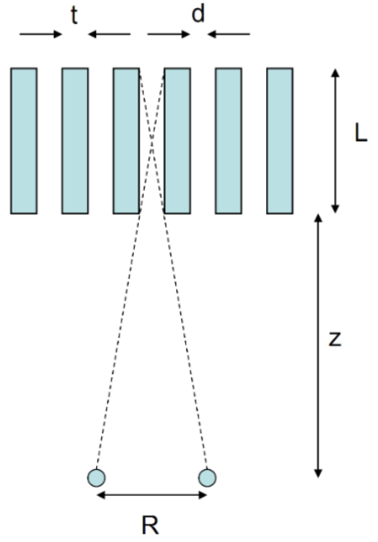


Fig. A parallel hole collimator, with parameters necessary to calculate the spatial resolution, R. The patient would be positioned at the bottom of the figure, with the gamma camera at the top.

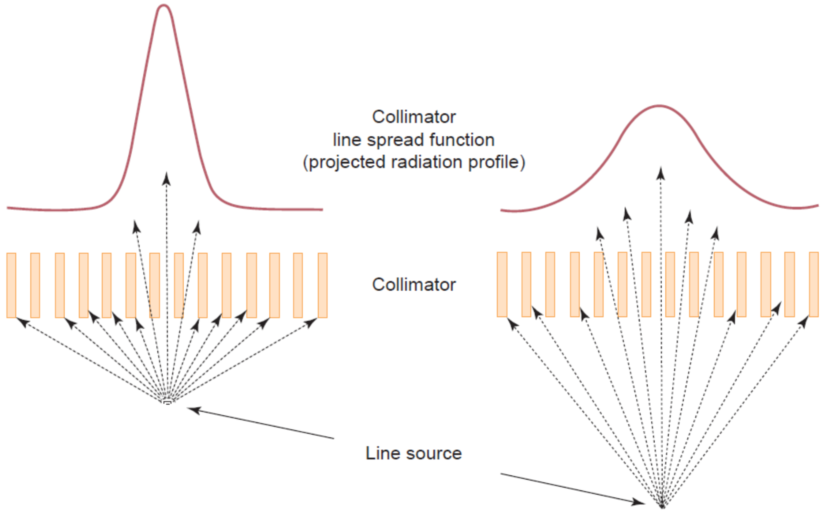


Fig. Line spread function (LSF) of a parallel-hole collimator as a function of source-to-collimator distance. The full-width-at-half-maximum (FWHM) of the LSF increases linearly with distance from the source to the collimator; however, the total area under the LSF (photon fluence through the collimator) decreases very little with source to collimator distance. (In both figures, the line source is seen “end-on.”)

Collimators

- **Parallel hole collimator:**
  - Low-energy all-purpose (LEAP): relatively large holes
  - Low-energy high-resolution (LEHR): smaller holes and longer septa
- **Slant hole collimator:** tilted parallel septa, primarily for breast and cardiac imaging
- **Converging collimator:** focused towards the body, used to magnify the image and increase the spatial resolution
- **Diverging collimator:** enables a larger FOV, for whole body imaging
- **Fan-beam collimator:**
  - Two dimensions of the collimator: parallel holes in head/foot direction, and converging holes in radial direction
  - Magnify image over a reduced FOV, for brain and heart studies
- **Pinhole collimator:** a single hole to image very small organs for significant magnification and higher spatial resolution.

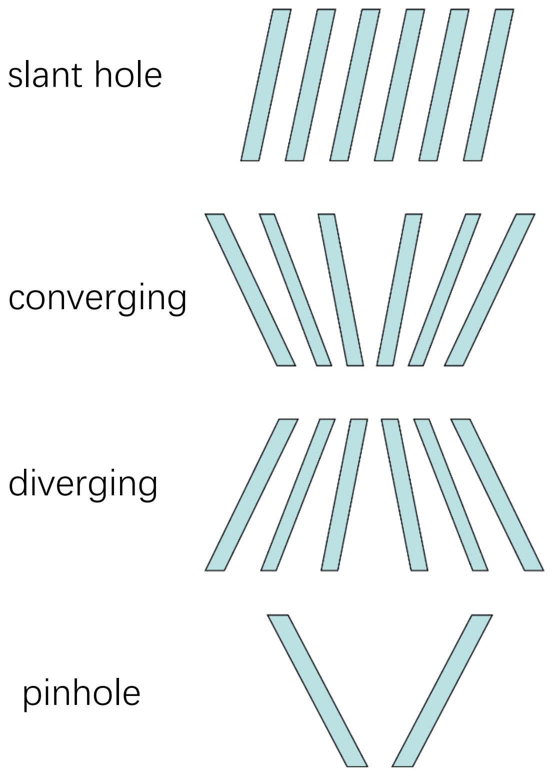
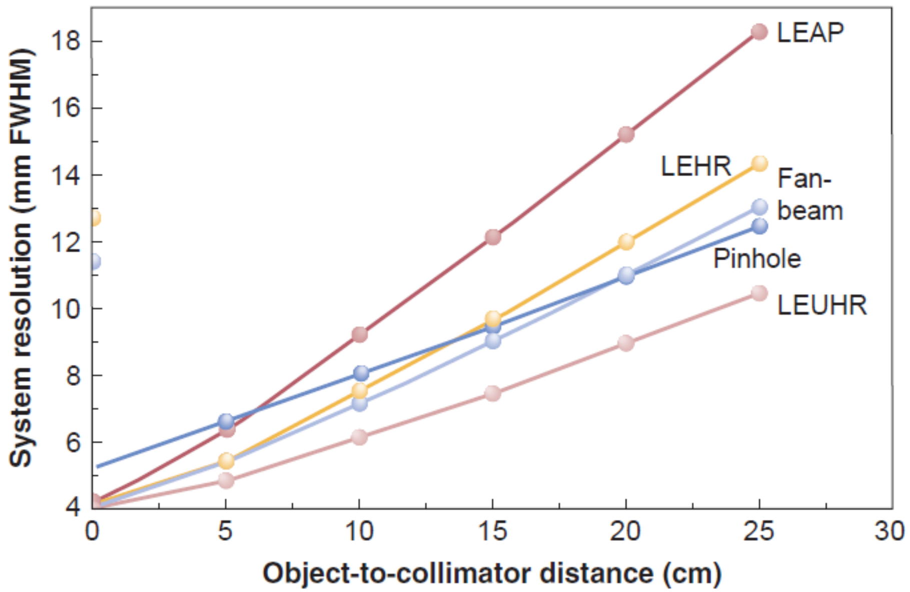


Fig. Four common types of collimator: from top-to-bottom, slanting hole, converging, diverging and pinhole.

Resolution and Efficiency of collimators



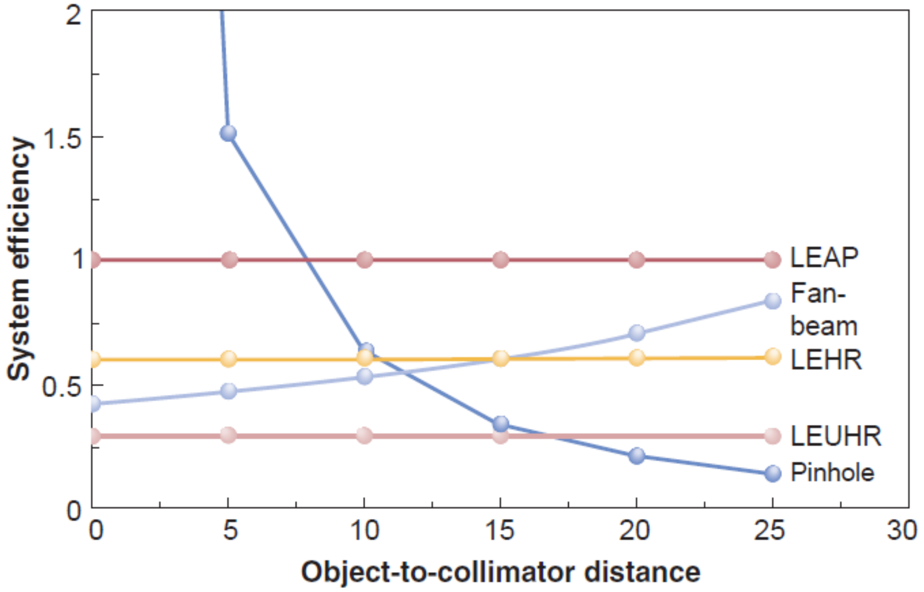


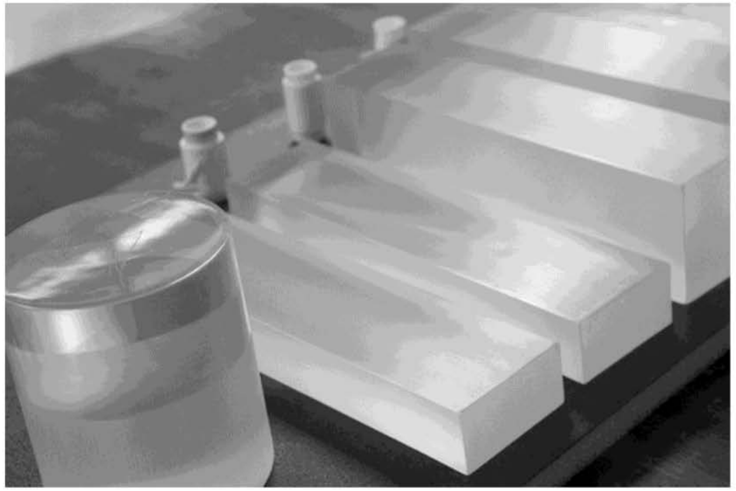
Fig. System spatial resolution (**top**) and efficiency (**bottom**) as a function of object-to-collimator distance (in cm). System resolutions for pinhole and fan-beam collimators are corrected for magnification. System efficiencies are relative to a low-energy, all-purpose (LEAP) parallel-hole collimator (system efficiency 340 cpm/μCi Tc-99m for a 0.95-cm- thick crystal). LEHR, low-energy, high-resolution parallel-hole collimator; LEUHR, low-energy, ultra-high-resolution parallel-hole collimator. (Data courtesy of the late William Guth of Siemens Medical Systems, Nuclear Medicine Group.) System spatial resolution (**top**) and efficiency (**bottom**) as a function of object-to-collimator distance (in cm). System resolutions for pinhole and fan-beam collimators are corrected for magnification. System efficiencies are relative to a low-energy, all-purpose (LEAP) parallel-hole collimator (system efficiency 340 cpm/μCi Tc-99m for a 0.95-cm- thick crystal). LEHR, low-energy, high-resolution parallel-hole collimator; LEUHR, low-energy, ultra-high-resolution parallel-hole collimator. (Data courtesy of the late William Guth of Siemens Medical Systems, Nuclear Medicine Group.)

Scintillation detector

The detector scintillation crystal (闪烁晶体探测器)

- High sensitivity, large detecting range of γ-ray energy
- Material: NaI(Tl). The advantages are:
  - High linear attenuation coefficient: 2.22/cm at 140KeV
  - Efficient: an efficiency of 15%. Produce one light photon per 30eV of energy absorbed, therefore the amount of light produced is directly proportional to the energy of the incident γ-ray; ~230ns de-excitation time
  - Transparent to its own light emission at 415nm
  - Large crystal can be grown easily and inexpensively: approximately 40-50cm diameter
  - 415nm wavelength is well matched to optimal performance of photomultiplier tubes
- The thickness of crystal:
  - Trade-off between spatial resolution and sensitivity
  - Generally used 1cm for various radiotracer detection

Crystals of NaI(Tl)



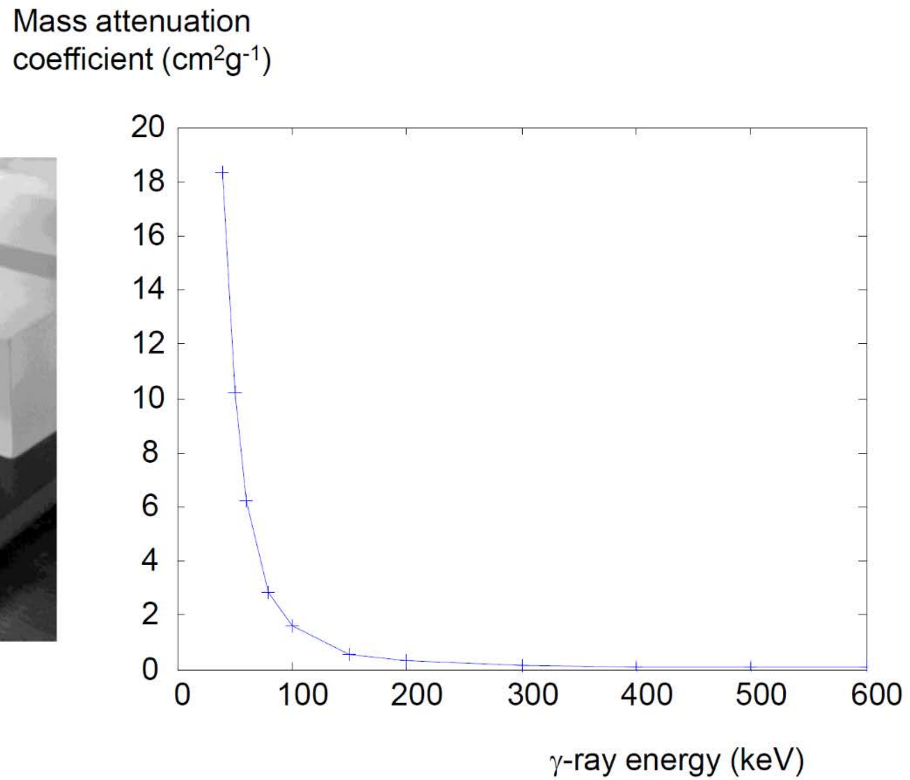


Fig. (left) Commercially produced crystals of NaI(Tl). (right) Plot of the mass attenuation coefficient for NaI(Tl) vs. γ-ray energy, showing an approximately exponential decrease with increasing energy.

PMT

Photon multiplier tubes (PMT, 光电倍增管)

- The produced photons with a very low energy (a few ev)
- PMT is used to amplify and convert low light signals to an electrical current which can be digitized
- The component of PMT
  - a photocathode(光电阴极): inside of the surface of transparent entrance window coated with material of CsSb.
  - First stage of dynode (倍增电极): the electron strikes and emit secondary electrons with amplification factor of 3-6 under a voltage of 300V
  - A series of 8-10 further accelerating dynodes at a voltage of 100V, produce 10^5-10^6 electrons for each initial photoelectron.
  - High voltage power supply in the range of 1-2kV
  - Vacuum glass tube to reduce attenuation of the electron

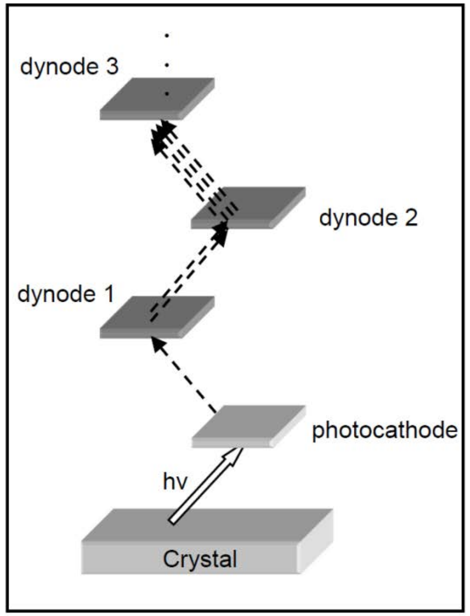


Fig. (left). The first three amplification stages in a PMT tube. (right) A commercial PMT.

PMT

Photon multiplier tubes (PMT, 光电倍增管)

- Diameter of 2-3cm
- Couple to the scintillation crystal with a thin optical coupling layer
- Typical arrays of 61, 75 or 91 PMTs
- Hexagonally-close-packed for the same distance between the centers of neighboring PMT which is important for the determination of spatial location of the scintillation event.
- Identical energy response for each PMT:
  - the variation in uniformity of up to 10% for planar gamma camera, less than 1% for SPECT
  - Calibration with sample of uniform and known radioactivity
  - Auto data correction can be applied to data

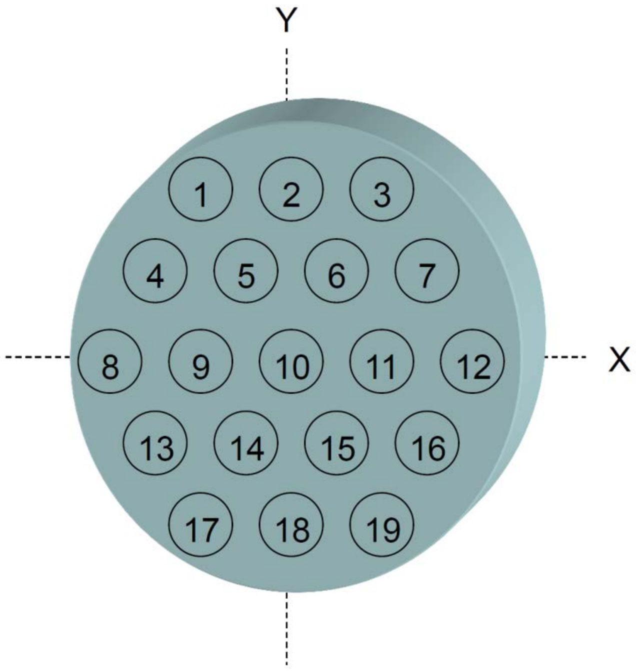


Fig. Nineteen PMTs in a hexagonal arrangement on an NaI(Tl) crystal

Anger position network

➢ The output of PMT is inversely proportional to the distance between the scintillation events and the PMT;

➢ The location of the scintillation event within the crystal can be estimated by comparing the magnitudes of current from all PMTs;

- **For analog camera: Anger logic circuit**

$$X^{+/-} = \frac{\sum_i x_i S_i}{\sum_i S_i} \qquad Y^{+/-} = \frac{\sum_i y_i S_i}{\sum_i S_i}$$

$$X = \frac{X^{+}-X^{-}}{X^{+}+X^{-}} \qquad Y = \frac{Y^{+}-Y^{-}}{Y^{+}+Y^{-}}$$

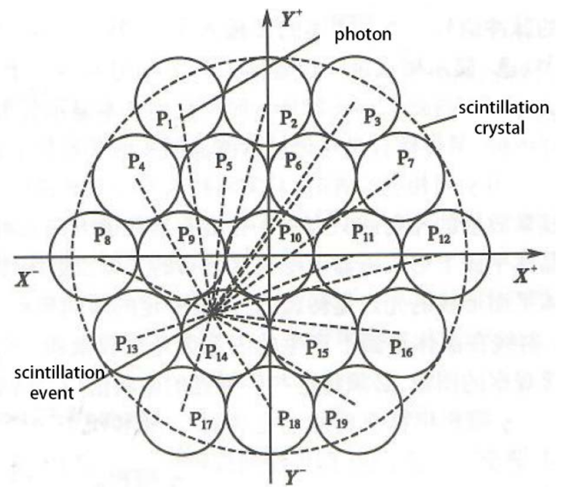
where (x_i, y_i): the position of the PMT

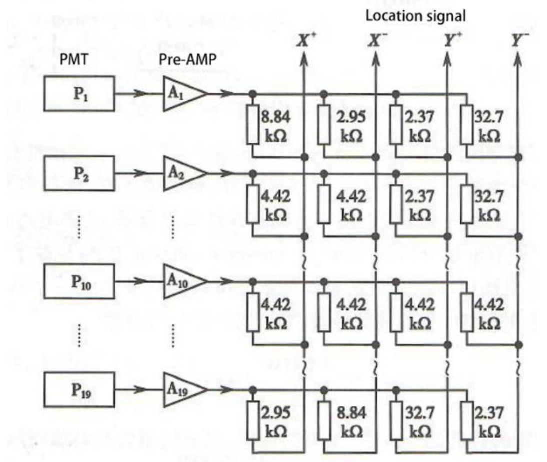
S_i : the integral of the PMT output over the scintillation duration

$X^{+/-}, Y^{+/-}$: sum output of all PMT

(X, Y) : the position of the scintillation event in the crystal

- **For digital camera:**
  - ✓ the position can be estimated directly from the stored currents for all the PMTs using computer algorithms;
  - ✓ Correction for non-uniform performance of each PMT can be built into the system





Pulse height analyzer

Pulse height analyzer (PHA, 脉冲幅度分析器)

- **Z** signal: summed signal of PMT output
  $\sum(X^{+}+X^{-}+Y^{+}+Y^{-})$
- Discriminate γ-ray by the magnitude of PMT output to determine which of the event corresponds to the γ-ray to be detected;
- Multiple-channel analyzer (MCA): digitize the signal and produce a pulse-height spectrum
- Apply upper and lower threshold values for accepting the γ-rays due to:
  - useful information provided by photon scattered small angles
  - Non-uniformity of crystal and PMT
- Energy resolution: FWHM of photopeak, typically 10% of photon energy, using 20% of photopeak in clinical scan

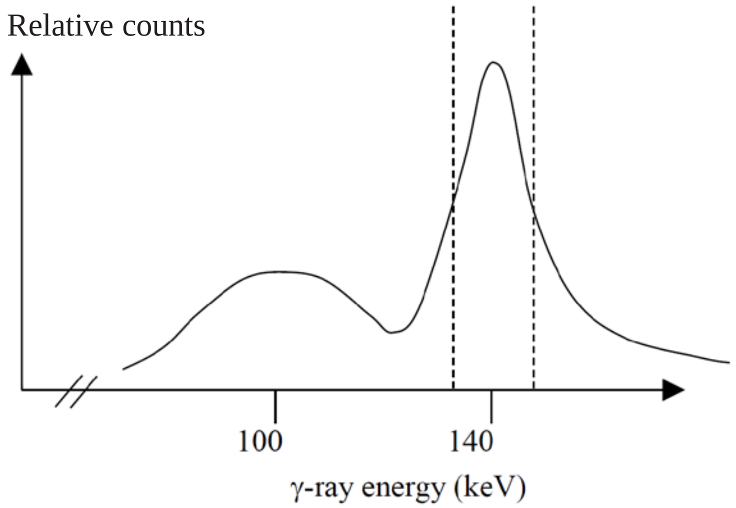


Fig. Readout from a multi-channel analyzer showing the number of counts recorded from a patient as a function of c-ray energy. The dashed lines show the energy resolution of the camera, defined as the FWHM of the main photopeak centered at 140 keV.

Instrument dead time

The limitation is due to the finite recovery and reset times required for various instrumentation in the gamma camera;

Two types of behaviors exhibited by the system components

- "Paralysable" component:
  - No response to a new event until a fixed time after the previous one
  - Dead time can be added even though the component is already in a non-responsive state;
  - Elongated dead time resulted from high radioactivity
  - The de-excitation of scintillation crystal
- "Non-paralysable" component:
  - No response for a set time irrespective of the level of radioactivity,
  - The processing time of Anger logic circuit and PHA;

Instrument dead time

- The dead time:

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

Where N: true count rate

n: measured count rate

- The maximum measurable count rate is $1/\tau$
- Correction of dead time based on
  - calibrations using phantom of known radioactivity; the experimental curve produced by imaging a source with known radioactivity
  - independent electronic measurements of individual components of the system;

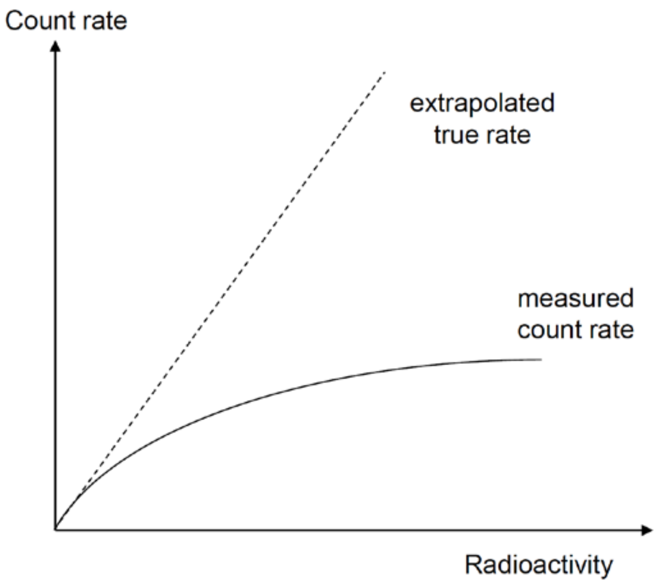


Fig. Correction curve used to account for dead time losses. At very low count rates, the true rate and measured count rate are linearly related. As the radioactivity increases, the measured count rate reaches an asymptotic value, well below the true value.

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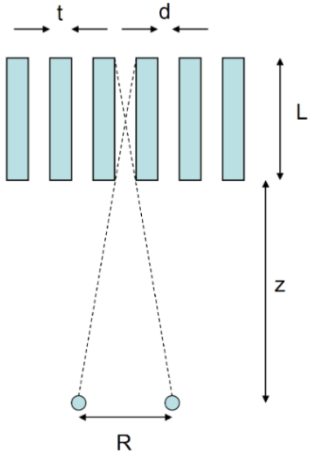
  - Photomultipliers tubes
  - Anger position network
  - Pulse height analyzer
  - Instrument dead time
- **Image characteristics**
  - **Influence of instrumentation**
  - **SNR**
  - **Spatial resolution**
  - **Contrast and contrast-to-noise**

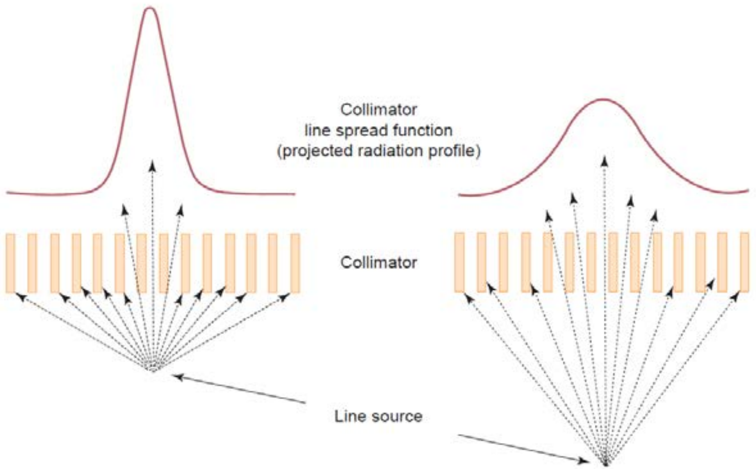
Influence from instrumentation

- From collimators
  - Reduce the number of recorded events (signal), therefore lead to lower SNR
  - Introduce significant depth-dependent broadening, therefore reduce the resolution

$$R = \frac{\mathrm{d}(\mathrm{L_{eff}} + z)}{\mathrm{L_{eff}}}$$

- Lowpass filtering
  - Required due to low SNR
  - Degrade the spatial resolution





SNR

Factors affecting the SNR includes

- Radioactivity of radiotracers
  - The amount of radiotracer administered
  - The amount of radiotracer that accumulates in the organ being imaged
  - The time after administering the radiotracer at which imaging begins
  - The total time of scan

  Note: the deeper the organ with the body, the lower SNR

- The intrinsic sensitivity of the gamma camera: mainly the collimator properties and geometry

- Post-acquisition imaging filtering

Resolution

- The spatial resolution of image is given as:

$$R_{\text{sys}} = \sqrt{R_{\text{gamma}}{}^2 + R_{\text{coll}}{}^2}$$

Where R_{gamma}: the intrinsic spatial resolution of the gamma camera

R_{coll}: the collimator resolution determined by the collimator geometry

- R_{gamma} is typical 3-5mm, and related to
  - uncertainty of the location of scintillation event
  - the thickness of the crystal
  - the intrinsic resolution of Anger position network (analog and digital)
- R_{coll} depends on
  - collimator types
  - The depth within the body of organs
- Typical system resolution
  - 1-2cm at large depth
  - 5-8mm close to the collimator surface

Contrast

- The theoretical image contrast is extremely high due to no background signal from tissue.
- Random background signals from Compton scattered γ-ray, especially close to the region where the radiotracer concentrate
- “Partial volume” effect: image blurring causes signal “bleed” from small area of high signal intensity to those where no radiotracer is present;
- For low-pass filter, there is a compromise between too strong a filter reducing the contrast and too weak a filter not reducing the noise sufficiently;

Comparison between radiography and Scintigraphy

	Planar Radiography (X-ray camera)	Planar Scintigraphy (Gamma camera)
Imaging modality	Transmission	Emission
Source	X-ray tube, outside of body	Radiotracer, in body
Collimator	Below X-ray tube	In front of detector
Detector	Photon energy	Photon count
Event location	FPD and TFT (DR)	PMT and position network
Intensity	High	Low
SNR	High	Low
Spatial resolution	High	Low, related to depth
Contrast	Low	High
Reaction with body	Photoelectric effect Compton scattering	