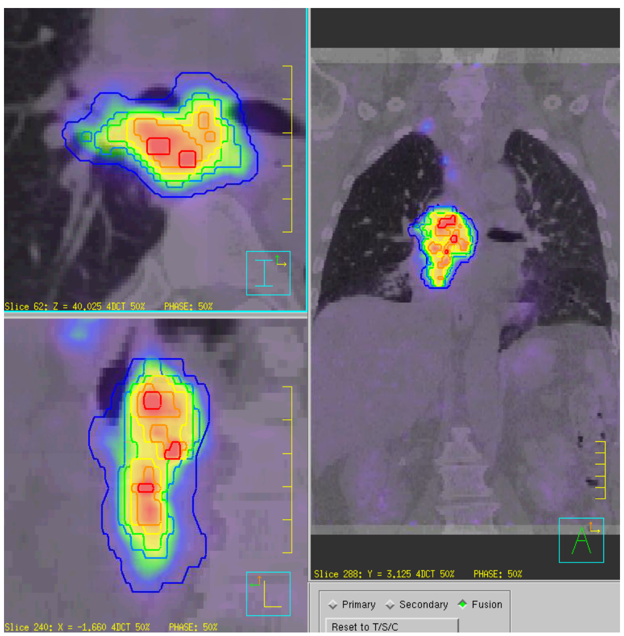
I was wondering about the list of "Preferable Characteristics for Nuclear Medicine Imaging Radionuclides" on slide 36 this week. What is chelation and why is that a desirable property for imaging isotopes?

chelation : attach a metal to a ring structure such that the “hide” or shield the metal atom from the biologically active sites of the molecule.

After our discussion about radiation dose in class I was wondering about how dose is managed across a patient's body for radiotherapy. If only one area of the body is receiving therapy, other areas of the body could be shielded with lead. Yet focused dose might damage tissues superficial or peripheral to a tumor. How is this addressed in treatment? If large areas of the body must be irradiated, what is done to account for body heterogeneity such as different layer thicknesses, densities, etc.?



Give radiation dose from different view angles.

Interest in proton therapy with Bragg peak and no radiation beyond Bragg peak.

If a lot of metastatic disease treat with chemotherapy or targeted radiation therapies.

Are longer lived radionuclides used for medical imaging? While the patient exposure is a concern, I would imagine biological half-life be manipulated through radiochemistry, but would there be any inherent value to longer lived radionuclides (energy peak(s) or cost reductions)?

Currently I-131 is used for medical imaging/therapy. It has an 8 day half-life.

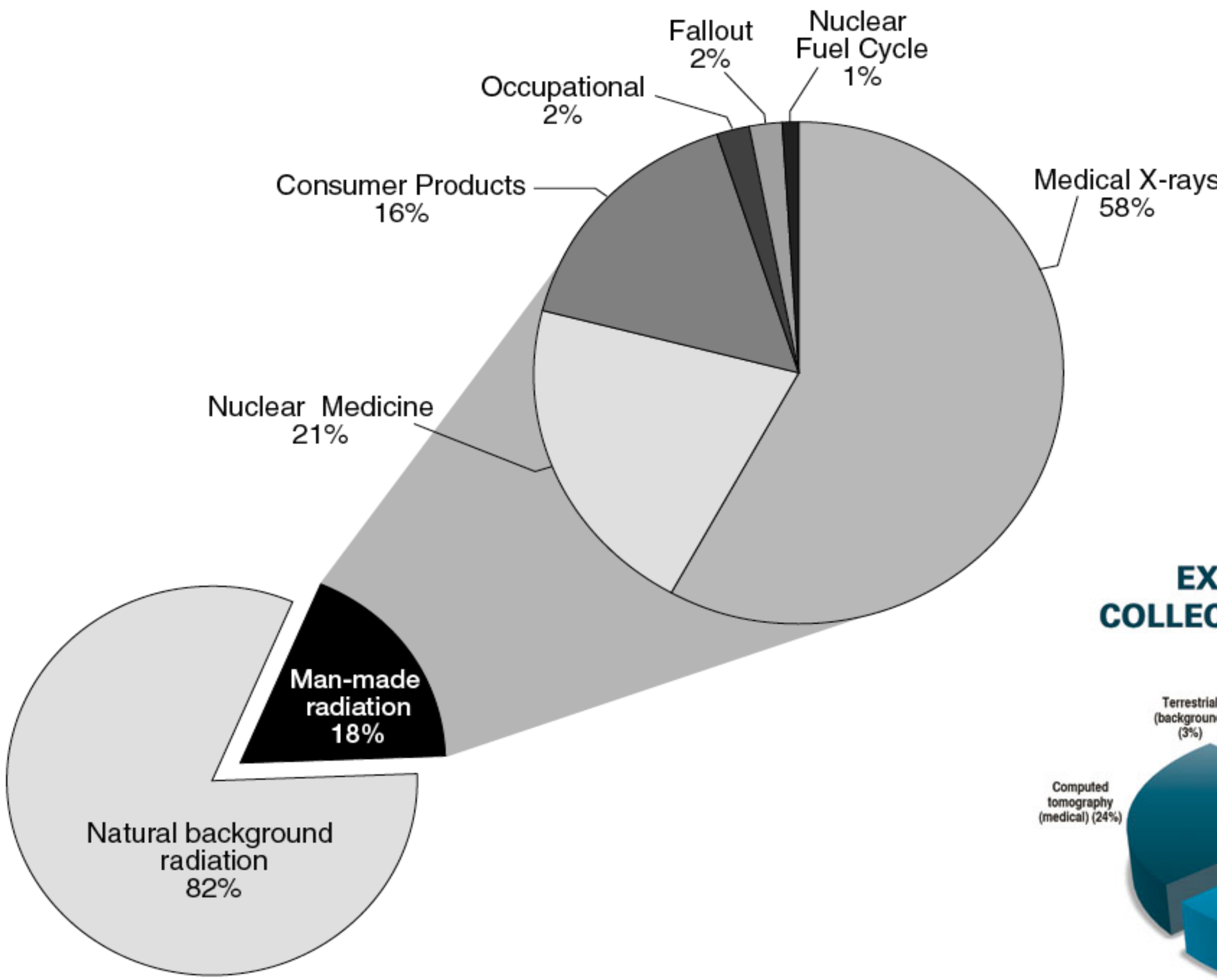
The issue with a long half life is obviously radiation exposure.

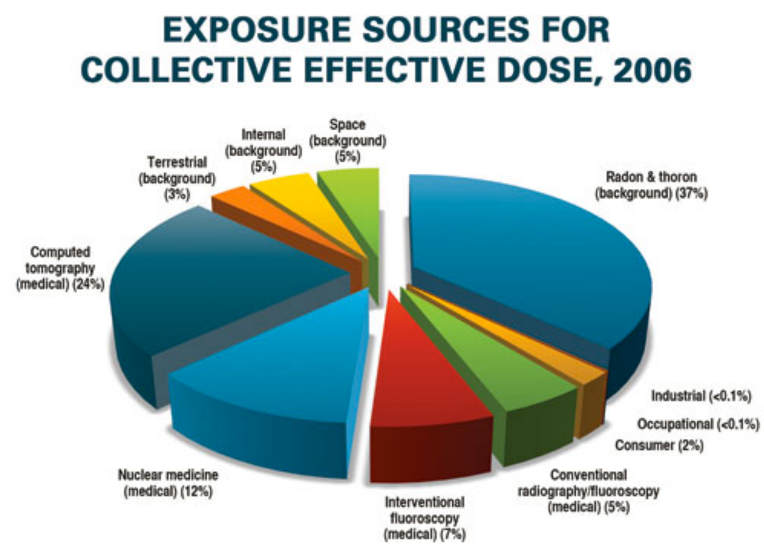
One research area where there is interest in long half-life radionuclides is in stem cell tracking. It is also of interest for some targeted antibody therapies to track the antibody over time. We are still talking about half lives of days, not months or years.

# Radiation Dosimetry

**a few beginning basics to a complex topic**

# Sources of Radiation Exposure in U.S.




This figure is based on data from “Ionizing Radiation Exposure of the Population of the United States”, *National Council on Radiation Protection and Measurements*, No.93, 1987.

# Dosimetry Descriptors - From Ionizing Radiation

*Exposure:* Amount of ionization of air caused by radioactive source
Charge per mass of air, **Coulomb/kg** = **3876 roentgens**
Can be measured directly
Does not account for biological effects

*Absorbed Dose:*
Energy per mass of tissue, Joules/kg = **gray (Gy)** = 100 rads
Usually calculated from exposure measurement
Does not account for biological effects

*Equivalent Dose:*
(Absorbed Dose) * radiation weighting factor ($w_R$ or $Q$ factor)
Also energy/mass, but units are **sieverts (Sv)** = 100 rem
Biological effects of absorbed dose depend on the type of radiation

*Effective Dose:*
Sum Over All Tissues[(Equivalent Dose$_T$) * tissue weighting factor ($w_T$)]
Also measured in **Sv**

.
The risk of cancer from a dose equivalent depends on the organ receiving the dose. The quantity "effective dose" is used to compare the risks when different organs are irradiated.

# Estimating Effective Dose

To go from absorbed dose (Gy) to equivalent dose (Sv), need:

**Radiation weighting factors**

| Type | $w_R$ |
|---|---|
| Photons | 1 |
| Electrons ($\beta$), muons | 1 |
| Neutrons (varies with energy) | 5-20 |
| Protons | 5 |
| alpha (a), heavy nuclei | 20 |

**For CT and PET, 1Gy = 1Sv**

International Commission on Radiological Protection, ICRP, Publ. 60, 1990 (www.icrp.org, *Annals of the ICRP*)

To go from Equivalent Dose (Sv) to Effective Dose (Sv), need:

**Tissue weighting factors**

| Tissue or organ | $w_T$ |
|---|---|
| Gonads | 0.20 |
| Bone marrow (red) | 0.12 |
| Colon | 0.12 |
| Lung | 0.12 |
| Stomach | 0.12 |
| Bladder | 0.05 |
| Breast | 0.05 |
| Liver | 0.05 |
| Esophagus | 0.05 |
| Thyroid | 0.05 |
| Skin | 0.01 |
| Bone surface | 0.01 |
| Remainder | 0.05 |
| **Total** | **1.00** |

$$w_T = \frac{D_{WB}(P)}{D_T(P)}$$

$D_{WB}(P)$= absorbed dose to the *whole body* that has probability $P$ of causing cancer

$D_T(P)$ = absorbed dose in a *single organ, T,* that has probability $P$ of causing cancer in that organ

# ALARA: As Low As Reasonable Achievable

**TABLE 23-18. NUCLEAR REGULATORY COMMISSION (NRC) REGULATORY REQUIREMENTS: MAXIMUM PERMISSIBLE DOSE EQUIVALENT LIMITS[a]**

| Limits | Maximum permissible annual dose limits | |
|---|---|---|
| | mSv | rem |
| Occupational limits | | |
| Total effective dose equivalent | 50 | 5 |
| Total dose equivalent to any individual organ (except lens of eye) | 500 | 50 |
| Dose equivalent to the lens of the eye | 150 | 15 |
| Dose equivalent to the skin or any extremity | 500 | 50 |
| Minor (<18 years old) | 10% of adult limits | 10% of adult limits |
| Dose to an embryo/fetus[b] | 5 in 9 months | 0.5 in 9 months |
| Nonoccupational (public limits) | | |
| Individual members of the public | 1.0/yr | 0.1/yr |
| Unrestricted area | 0.02 in any 1 hr[c] | 0.002 in any 1 hr[c] |

[a]These limits are exclusive of natural background and any dose the individual has received for medical purposes; inclusive of internal committed dose equivalent and external effective dose equivalent (i.e., total effective dose equivalent).
[b]Applies only to conceptus of a worker who declares her pregnancy. If the limit exceeds 4.5 mSv (450 mrem) at declaration, conceptus dose for remainder of gestation is not to exceed 0.5 mSv (50 mrem).
[c]This means the dose to an area (irrespective of occupancy) shall not exceed 0.02 mSv (2 mrem) in any 1 hour. This is not a restriction of instantaneous dose rate to 0.02 mSv/hr (2 mrem/hr).

**Shielding**

**Distance**

**Exposure time**

# Average Dose Equivalent

**TABLE 23-3. AVERAGE ANNUAL OCCUPATIONAL EFFECTIVE DOSE EQUIVALENT IN THE UNITED STATES**

| | Average annual total effective dose equivalent | |
|---|---|---|
| **Occupational category** | **mSv** | **mrem** |
| Uranium miners[a] | 12.0 | 1,200 |
| Nuclear power operations[b] | 6.0 | 600 |
| Airline crews | 1.7 | 170 |
| Diagnostic radiology and nuclear medicine techs | 1.0 | 100 |
| Radiologists | 0.7 | 70 |

Adapted for measurably exposed personnel from National Council on Radiation Protection and Measurements. *Exposure of the U.S. population from occupational radiation.* NCRP report no. 101. Bethesda, MD: National Council on Radiation Protection and Measurements, 1989.
[a]Includes 10 mSv (1 rem) from high LET ($\alpha$) radiation.
[b]Includes 0.5 mSv (50 mrem) from high LET ($\alpha$) radiation.
LET, linear energy transfer.

Bushberg et al, The Essential Physics of Medical Imaging, Lippencott, Williams & Wilkins, Philadephia, 2002.

**TABLE 9.1.** *Characteristics of common radionuclides*

| Nuclide | Photons (keV) | Production mode | Decay mode | Half-life ($T_{1/2}$) |
|---|---|---|---|---|
| $^{67}Ga$ | 93, 185, 296, 388 | Cyclotron | EC | 78 hr |
| $^{99m}Tc$ | 140 | Generator | IT | 6 hr |
| $^{111}In$ | 173, 247 | Cyclotron | EC | 68 hr |
| $^{123}I$ | 159 | Cyclotron | EC | 13 hr |
| $^{125}I$ | 27, 36 | Reactor | EC | 60 d |
| $^{131}I$ | 364 | Fission product | β | 8 d |
| $^{133}Xe$ | 80 | Fission product | β | 5.3 d |
| $^{201}Tl$ | 70, 167 | Cyclotron | EC | 73 hr |

β, beta decay; EC, electron capture; IT, isomeric transition.

Some Reactor-produced Radionuclides Used in Nuclear Medicine and Radiotracer Kinetics

| Radionuclide | Decay Mode | Production Reaction | Natural Abundance of Target Isotope (%) | $\sigma_c$ (b)* |
|---|---|---|---|---|
| $^{14}C$ | $\beta^-$ | $^{14}N(n,p)^{14}C$ | 99.6 | 1.81 |
| $^{24}Na$ | $(\beta^-,\gamma)$ | $^{23}Na(n,\gamma)^{24}Na$ | 100 | 0.53 |
| $^{32}P$ | $\beta^-$ | $^{31}P(n,\gamma)^{32}P$ | 100 | 0.19 |
| | | $^{32}S(n,p)^{32}P$ | 95.0 | — |
| $^{35}S$ | $\beta^-$ | $^{35}Cl(n,p)^{35}S$ | 75.5 | — |
| $^{42}K$ | $(\beta^-,\gamma)$ | $^{41}K(n,\gamma)^{42}K$ | 6.8 | 1.2 |
| $^{51}Cr$ | $(EC,\gamma)$ | $^{50}Cr(n,\gamma)^{51}Cr$ | 4.3 | 17 |
| $^{59}Fe$ | $(\beta^-,\gamma)$ | $^{58}Fe(n,\gamma)^{59}Fe$ | 0.3 | 1.1 |
| $^{75}Se$ | $(EC,\gamma)$ | $^{74}Se(n,\gamma)^{75}Se$ | 0.9 | 30 |
| $^{125}I$ | $(EC,\gamma)$ | $^{124}Xe(n,\gamma)^{125}Xe \xrightarrow{EC} {}^{125}I$ | 0.1 | 110 |
| $^{131}I$ | $(\beta^-,\gamma)$ | $^{130}Te(n,\gamma)^{131}Te \xrightarrow{\beta^-} {}^{131}I$ | 34.5 | 0.24 |

*Thermal neutron capture cross-section, in barns, for $(n,\gamma)$ reactions (see Section D.1).

# Properties of Gamma Rays and Beta Rays

Gamma Rays

massless photons travel potentially long distances in body

➔ emitted with single energy (mono-energetic, allows energy discrimination)

➔ penetration is exponential: $N=N_0e^{-\mu(E,Z,r,\text{interaction})*x}$

➔ typical ~ cm-to-m penetration, no limits to penetration depth

➔ difficult to collimate – requires high Z &/or high density material (e.g Pb, W)

Beta Rays (e- & e+)

charged particles with mass undergo many interactions in body

➔ emitted with continuous energy distribution (energy discrimination not effective)

➔ no analytical rule for penetration depth (between exp.&linear)

➔ typical ~ mm penetration, maximum penetration depends on particle E

➔ easy to collimate

# Gamma Cameras: Components and Systems

Robert Miyaoka, Ph.D.
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# Interactions of high energy photons with matter

Photoelectric effect

Compton scattering

Pair production

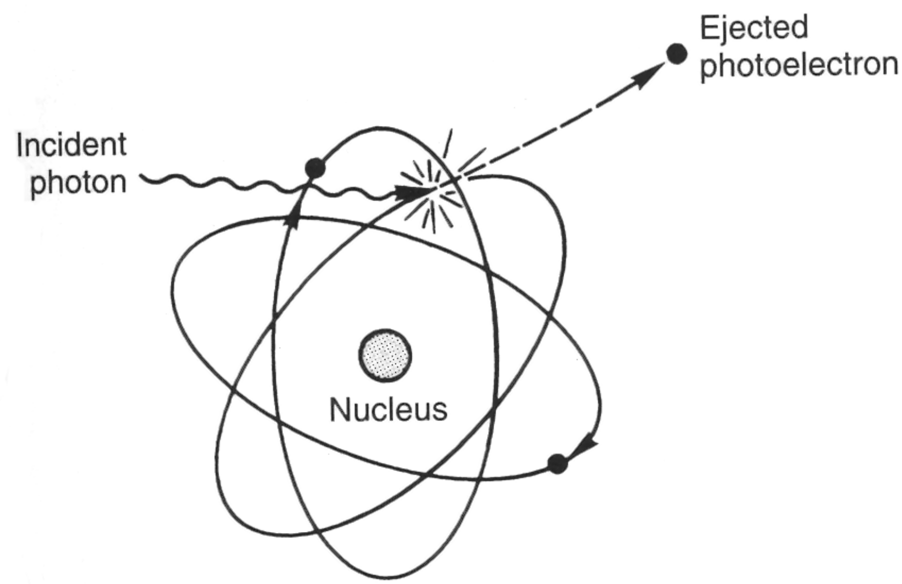
Coherent (Rayleigh) scattering

# Interactions of photons with matter
## (photoelectric effect)

An atomic absorption process in which an atom absorbs all the energy of an incident photon.

$$PE \propto \frac{Z^3}{E^3 \rho}$$

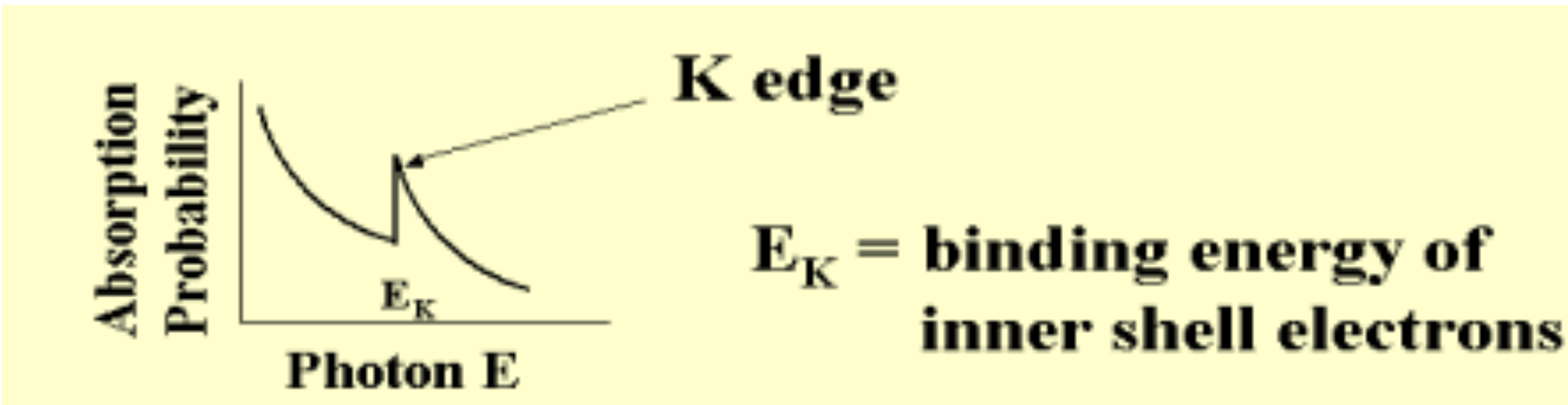
$Z$ is atomic number of the material, $E$ is energy of the incident photon, and $\rho$ is the density of the material.




From: Physics in Nuclear Medicine (Cherry, Sorenson and Phelps)

# Interactions of photons with matter
## (photoelectric effect)

Photons are preferentially absorbed by more tightly bound electrons.

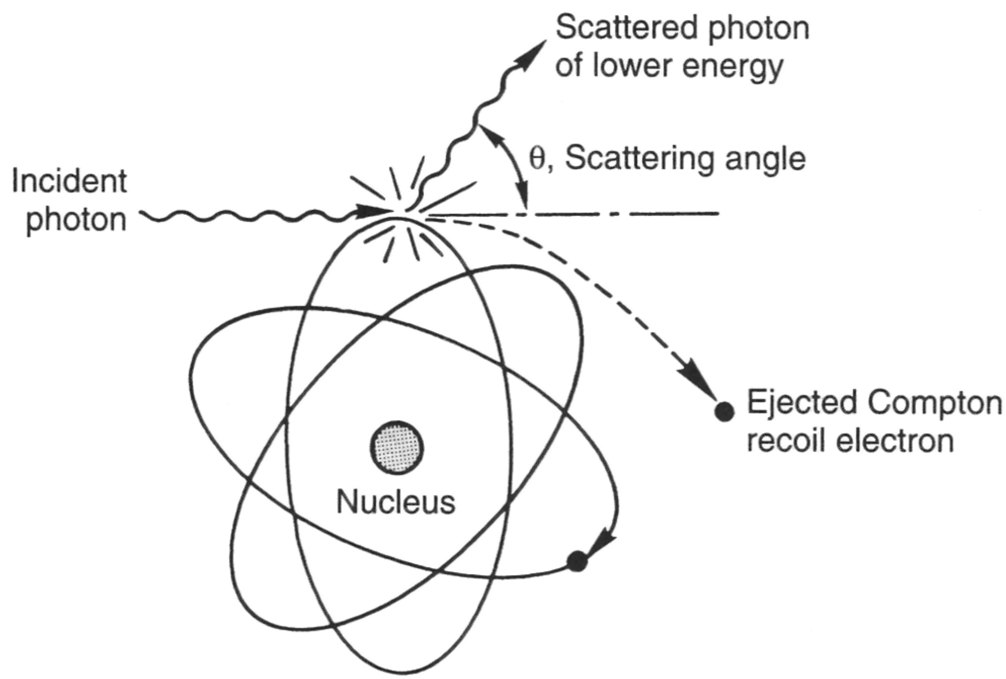



Photoelectric absorption involving K-shell electrons cannot occur until the photon energy exceeds the K-shell binding energy.

From: Society of Nuclear Medicine: Basic Science of Nuclear Medicine CD

# Interactions of photons with matter
## (Compton scatter)

Collision between a photon and a loosely bound outer shell orbital electron. Interaction looks like a collision between the photon and a "free" electron.




From: Physics in Nuclear Medicine (Cherry, Sorenson and Phelps)

# Interactions of photons with matter
## (Compton scatter)

The probability of Compton scatter is a slowly varying function of energy. It is proportional to the density of the material (*r*) but independent of Z.

$$CS \propto \rho$$

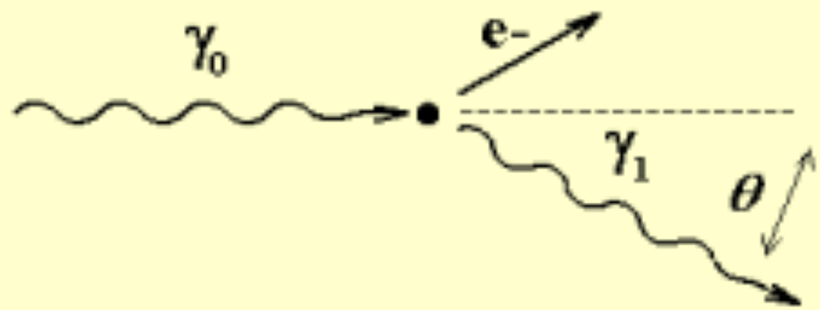
*r* is the density of the material.

As energy is increased scatter is forward peaked.

# Interactions of photons with matter
## (Compton scatter)

The scattering angle is determined by the amount of energy transferred in the collision.

$$E_1 = \frac{E_0}{1 + \frac{E_0}{m_e c^2}(1 - \cos\theta)}$$




**$E_0$ is original photon energy**
**$E_1$ is scattered photon's energy**
**$m_e$ is the electron mass ($m_e c^2$=511 keV)**
**$\theta$ is the scattering angle ($\theta$=0 is no scatter)**

From: Society of Nuclear Medicine: Basic Science of Nuclear Medicine CD

# Polar plot of relative frequency of scattering angles

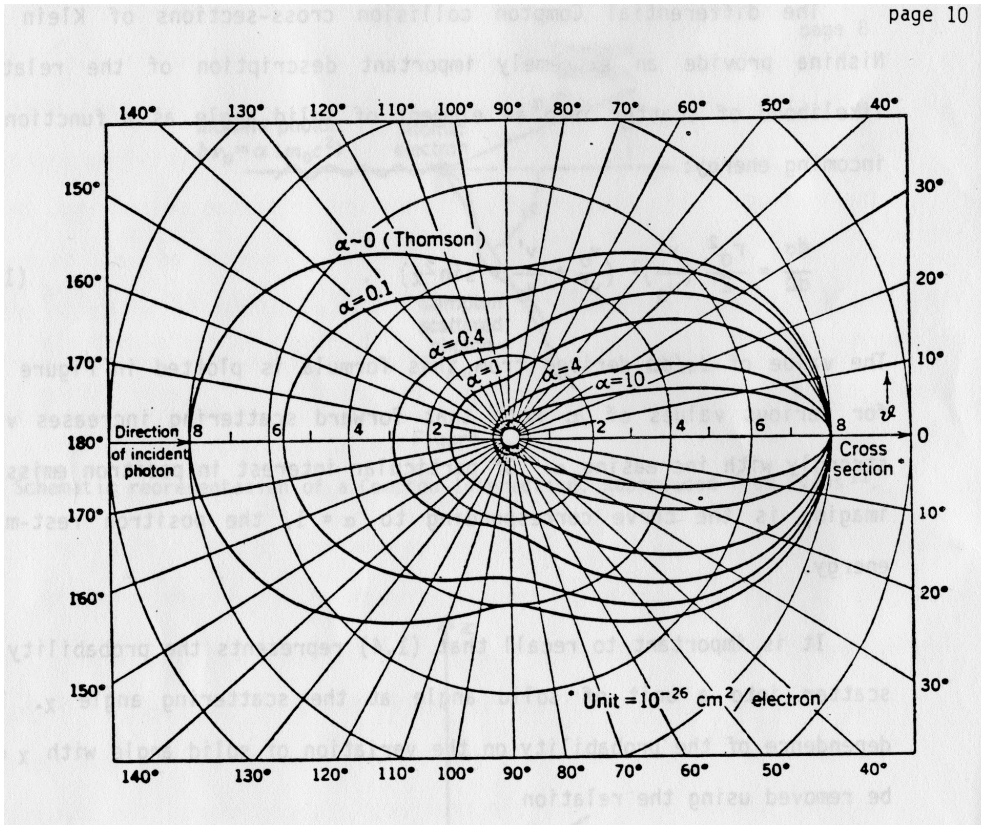



Figure 1.5

Polar plot of the relative number of photons scattered into a unit of solid angle in the direction of the scattering angle χ. Reproduced from Evans[12].

From: Bruno, UC Berkeley masters thesis

# Interactions of photons with matter
## (pair production)

Pair production occurs when a photon interacts with the electric field of a charged particle. Usually the interaction is with an atomic nucleus but occasionally it is with an electron.

Photon energy is converted into an electron-positron pair and kinetic energy. Initial photon must have an energy of greater than 1.022 MeV.

Positron will eventually interact with a free electron and produce a pair of 511 keV annihilation photons.

# Interactions of photons with matter
## (Coherent or Rayleigh scatter)

Scattering interactions that occur between a photon and an atom as a whole.

Because of the great mass of an atom very little recoil energy is absorbed by the atom.  The photon is therefore deflected with essentially no loss of energy.

Coherent scattering is only important at energies <50 keV.

# Attenuation

Under conditions of narrow beam geometry the transmission of a monoenergetic photon beam through an absorber is described by an exponential equation:

$$I(x) = I(0)e^{-\mu x} \quad ,$$

where *I(0)* is the initial beam intensity, *I(x)* is the beam intensity transmitted through a thickness *x* of absorber, and $\mu$ is the total linear attenuation coefficient of the absorber at the photon energy of interest.

The linear attenuation coefficient is expressed in units of $cm^{-1}$.

Attenuation is not the same as absorption.

# Attenuation coefficients

There are three basic components to the linear attenuation coefficient: $\mu_t$ due to the photoelectric effect; $\mu_s$ due to Compton scattering; and $\mu_k$ due to pair production. The exponential equation can also be written as:

$$I(x) = I(0)e^{-(\mu_\tau + \mu_\sigma + \mu_\kappa)x}$$

or

$$I(x) = I(0)e^{-\mu_\tau x}e^{-\mu_\sigma x}e^{-\mu_\kappa x}$$

# Relative magnitudes

- **Dominating effect depends on photon energy:**
  - **low E: photoelectric absorption**
  - **medium E: Compton scattering**
  - **high E: pair production**
- **Crossover points depend on material:**

**lead:**

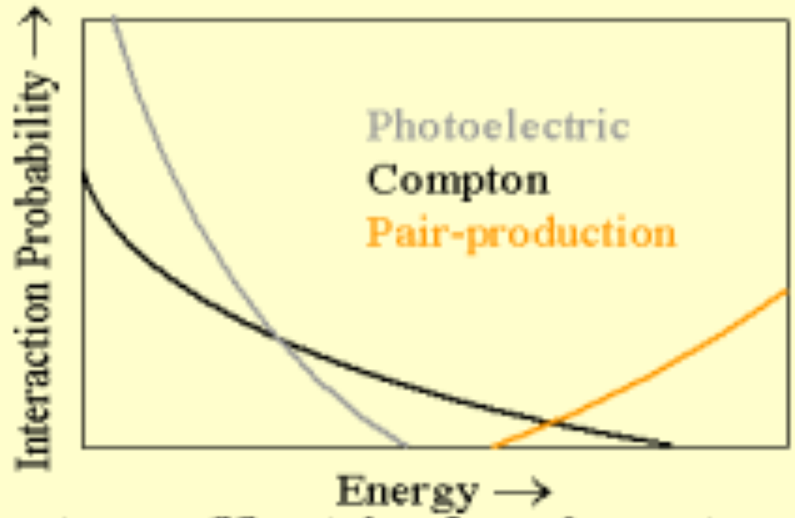
**PE-C ~ 500 keV**

**C-PP ~ 5 MeV**

**water:**

**PE-C ~ 50 keV**

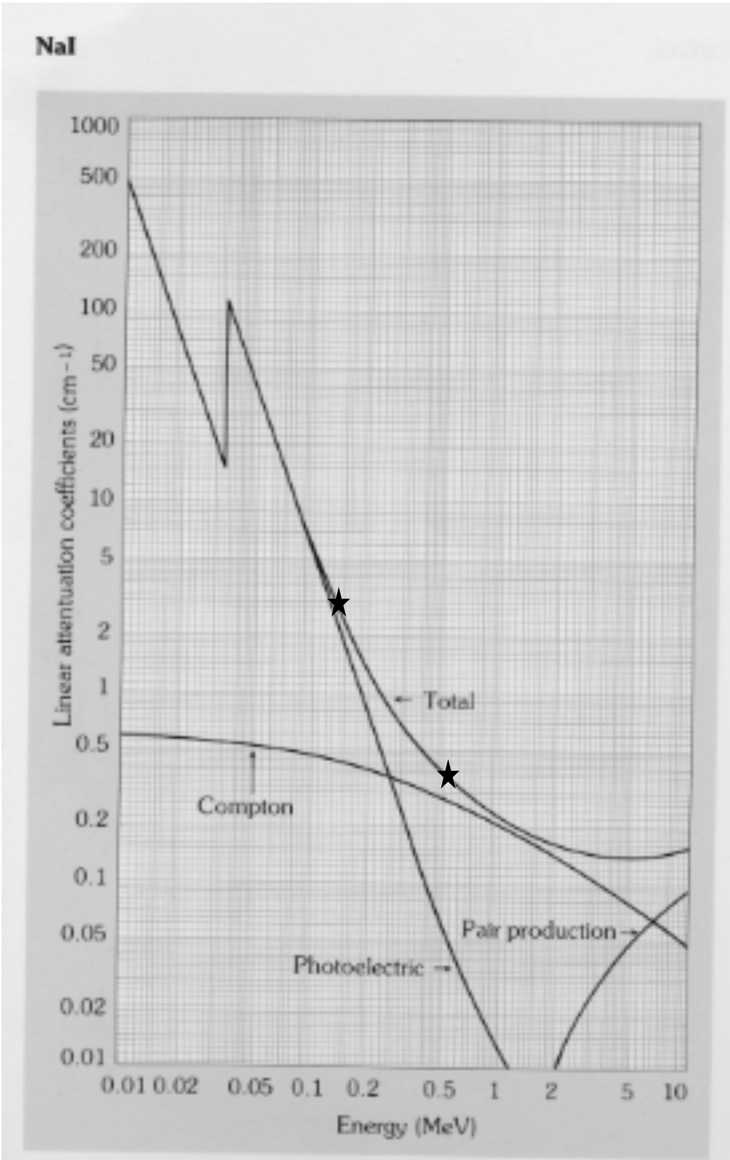
**C-PP ~ 20 MeV**

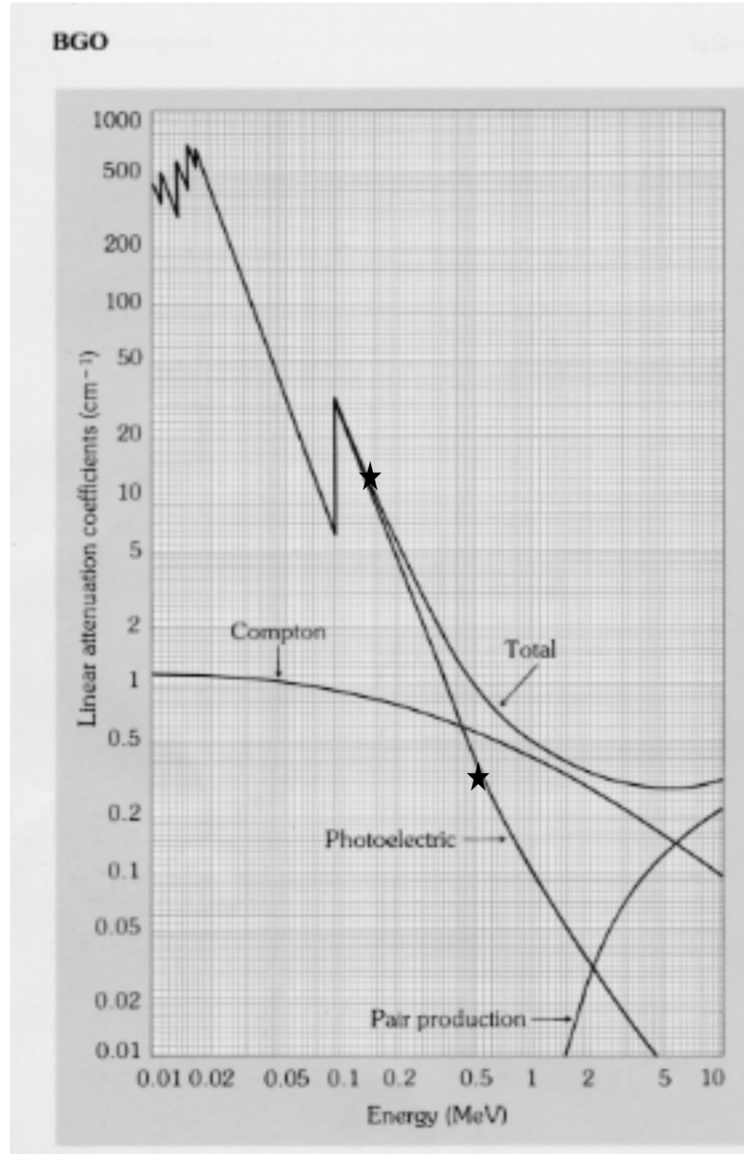



- **For NM energies, Compton effect is dominant**

From: Society of Nuclear Medicine: Basic Science of Nuclear Medicine CD

# NaI(Tl) and BGO linear attenuation coefficients

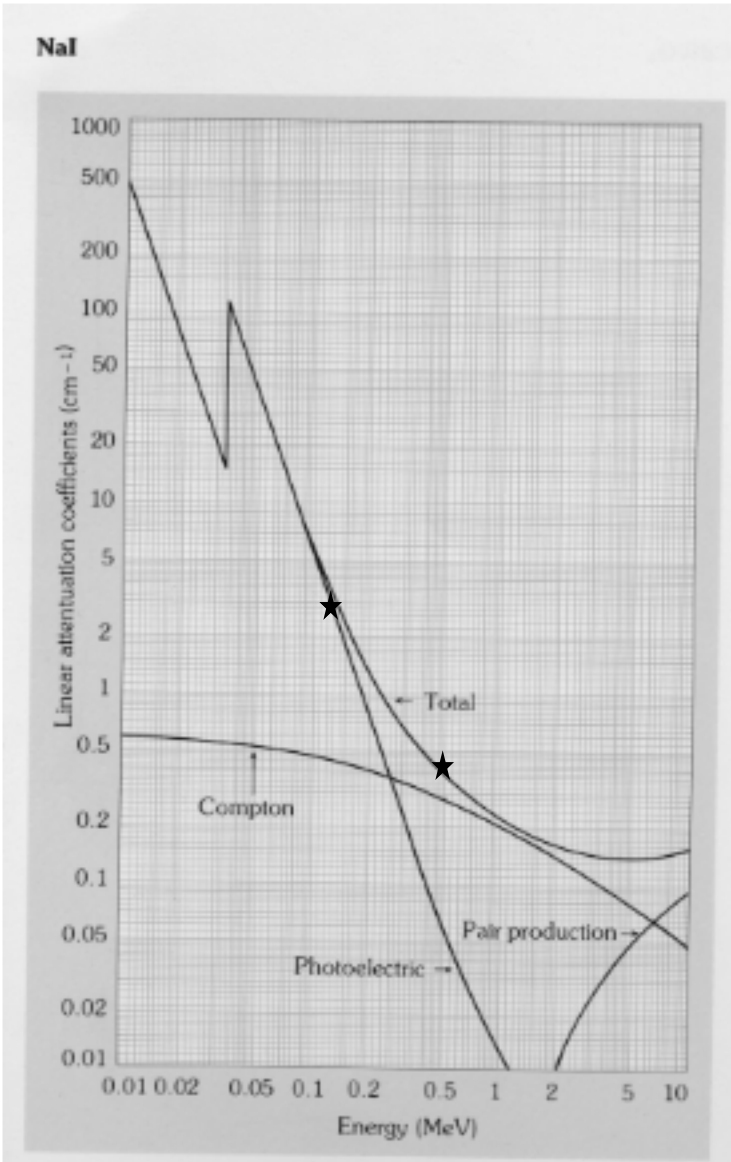


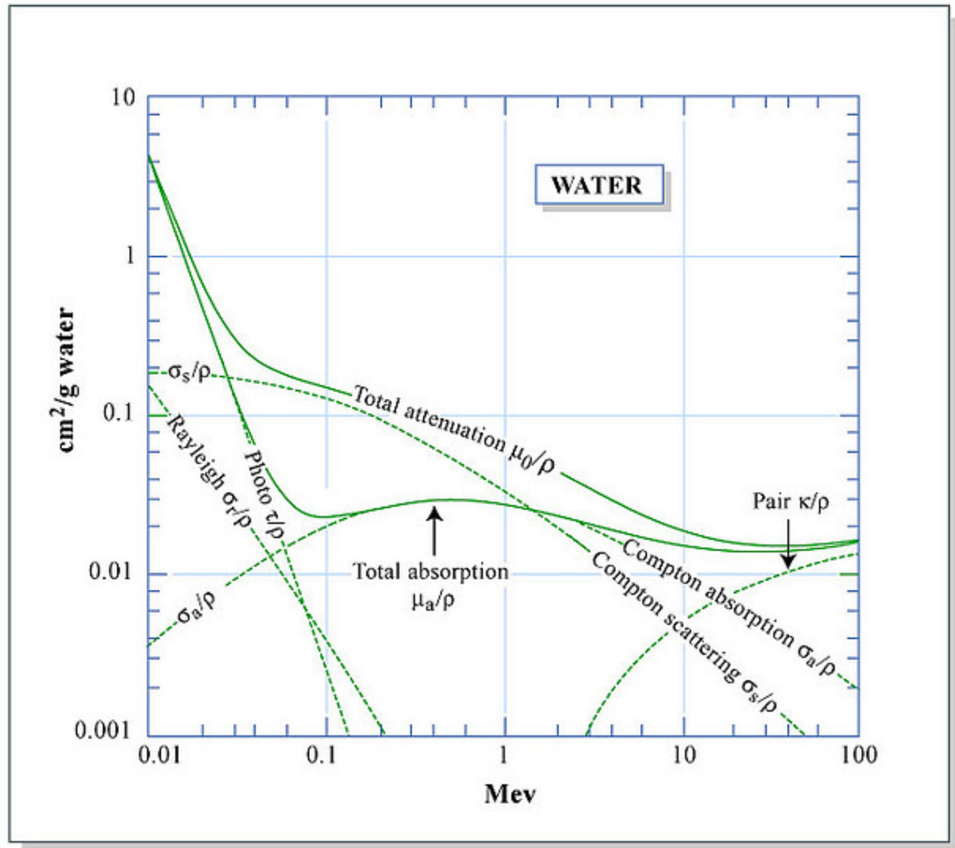




From: Bicron (Harshaw) Scintillator catalog

Robert Miyaoka, Ph.D. May 21, 2013

# NaI(Tl) and Water linear attenuation coefficients




MIT OpenCourseWare

From: Bicron (Harshaw) Scintillator catalog

# Attenuation Coefficient

Linear attenuation coefficient $\mu_l$

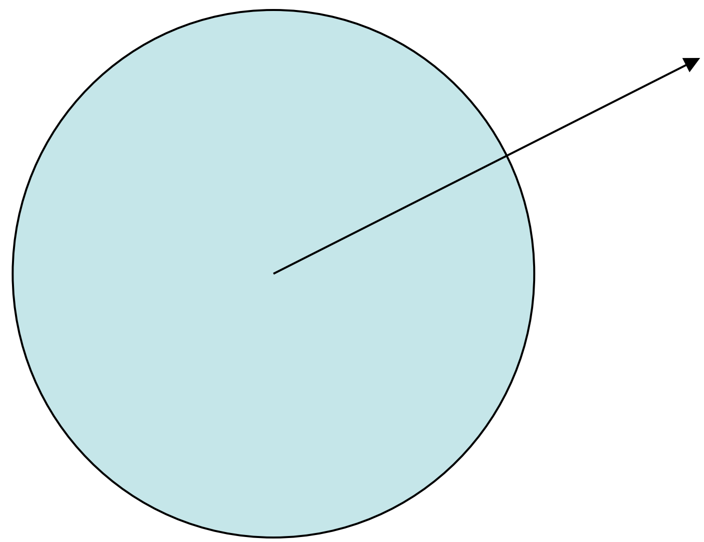
depends on photon energy
depends on material composition
depends on material density
dimensions are 1/length (e.g., 1/cm, $cm^{-1}$)

Mass attenuation coefficient $\mu_m$

$\mu_m = \mu_l / \rho$ ($\rho$ = density of material yielding $\mu_l$ )
does not depend on material density
dimensions are $length^2$/mass (e.g., $cm^2$/g)

# Example Calculation

What fraction of 140 keV photons will escape unscattered from the middle of a 30 cm cylinder?

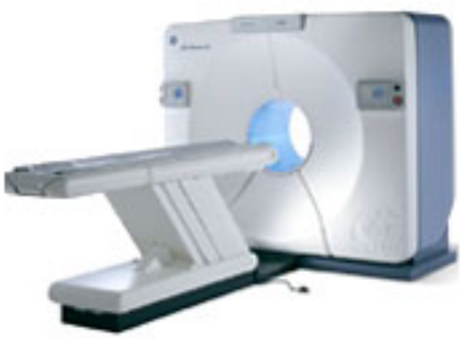


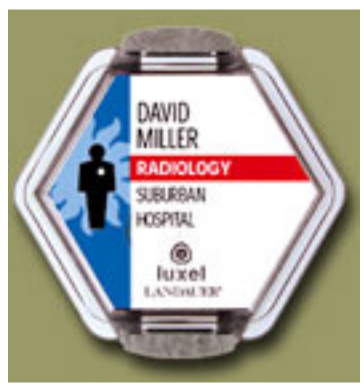
The photons must travel through 15 cm of water.

$I/I_0 = e^{-\mu d} = e^{-(0.15/cm)(15cm)} = 0.105 = 10.5\%$

Robert Miyaoka, Ph.D.  May 21, 2013

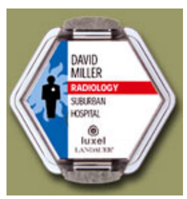
# Radiation Detectors




# Types of Radiation Detectors

- Gas-filled detectors
- Solid-state (semiconductor) detectors
- Organic liquid scintillators
- Film

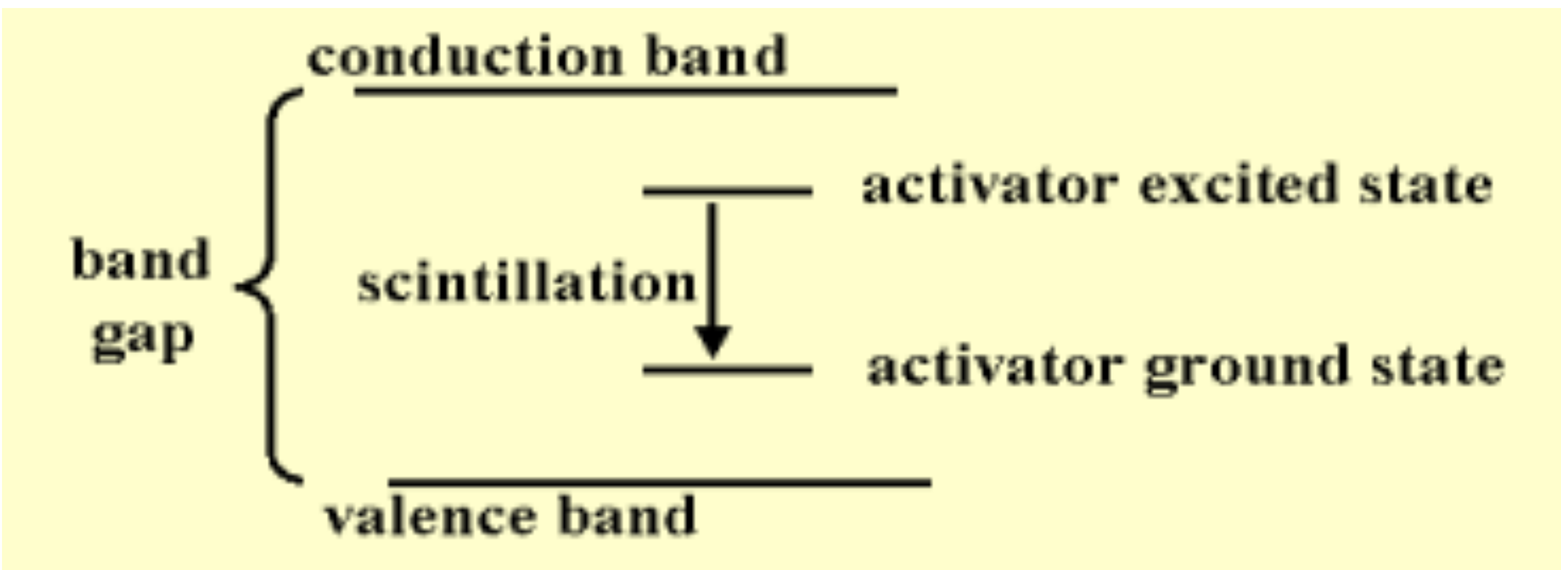



- **Inorganic scintillators**

Why are inorganic scintillators mostly used for gamma cameras?

# Inorgranic Scintillators

- crystalline solids
- scintillate because of characteristics of crystal structure
- impurities often required for scintillation properties

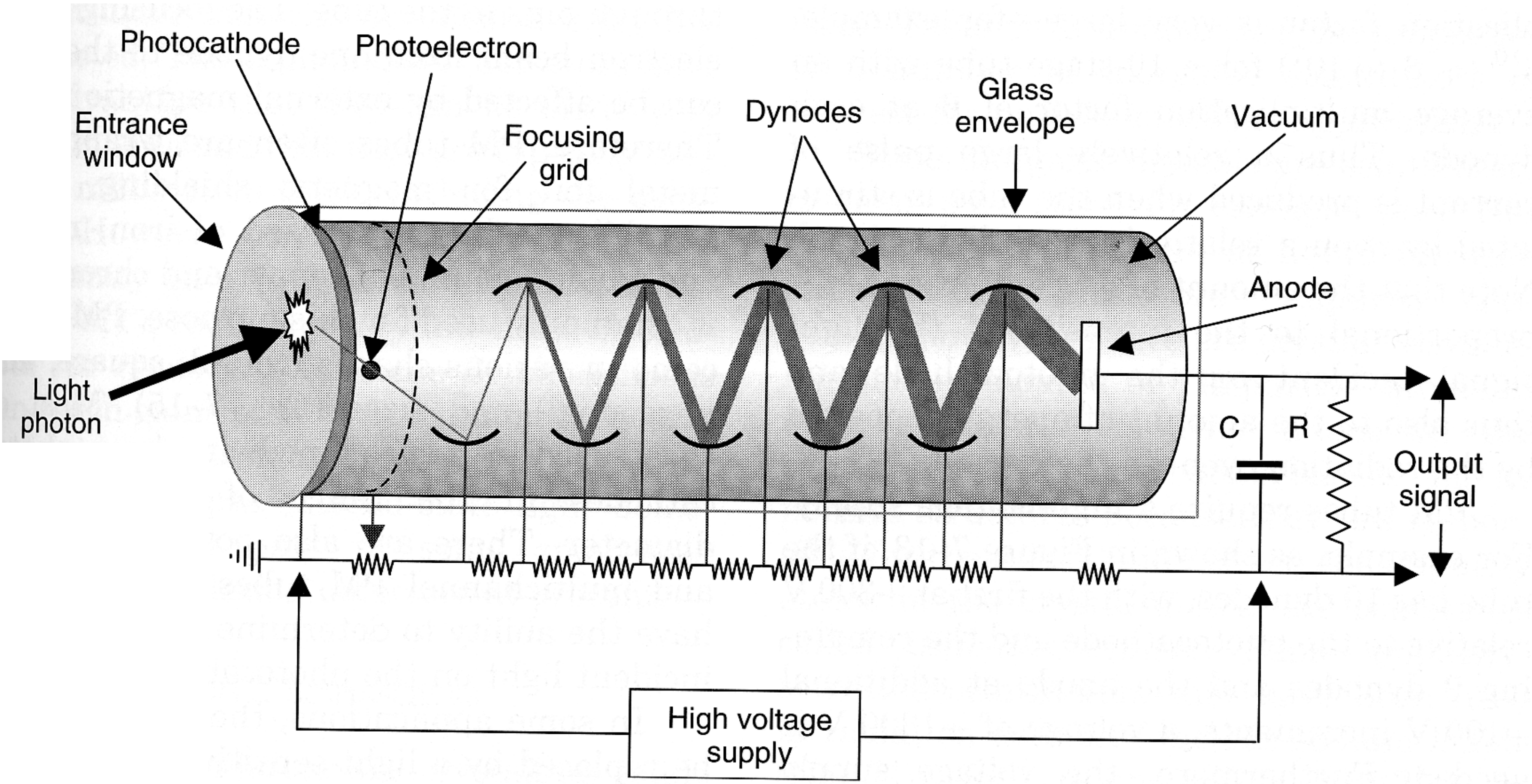



From: Society of Nuclear Medicine: Basic Science of Nuclear Medicine CD

# Inorganic scintillators

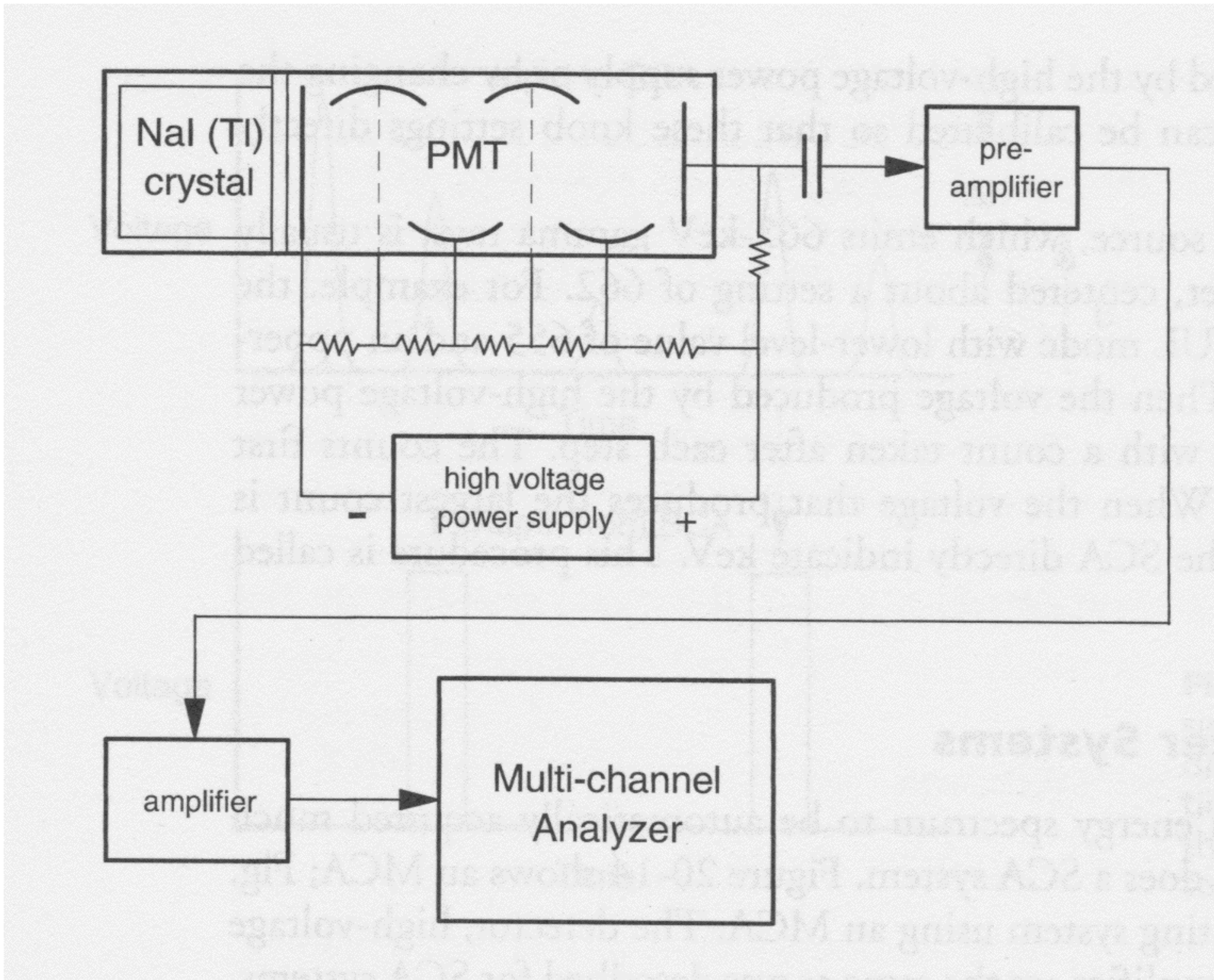
| | NaI(Tl) | BGO | LSO(Ce) | GSO(Ce) | |
|---|---|---|---|---|---|
| Density (gm/cm$^3$) | 3.67 | 7.13 | 7.4 | 6.71 | Sensitivity |
| Effective Atomic Number | 51 | 75 | 66 | 59 | |
| Attenuation Coefficient (@ 511 keV, cm$^{-1}$) | 0.34 | 0.955 | 0.833 | 0.674 | |
| Light Output (photons/Mev) | 40K | ~8K | ~30K | ~20K | Eng and spat res |
| Decay Time | 230 ns | 300 ns | 12 ns<br>40 ns | 60 ns | Counting speed |
| Wavelength | 410 nm | 480 nm | 420 nm | 430 nm | Photo-sensor / Cost |
| Index of Refraction | 1.85 | 2.15 | 1.82 | 1.85 | |
| Hygroscopy | yes | no | no | no | |
| Rugged | no | yes | yes | no | |

# Photomultiplier Tube



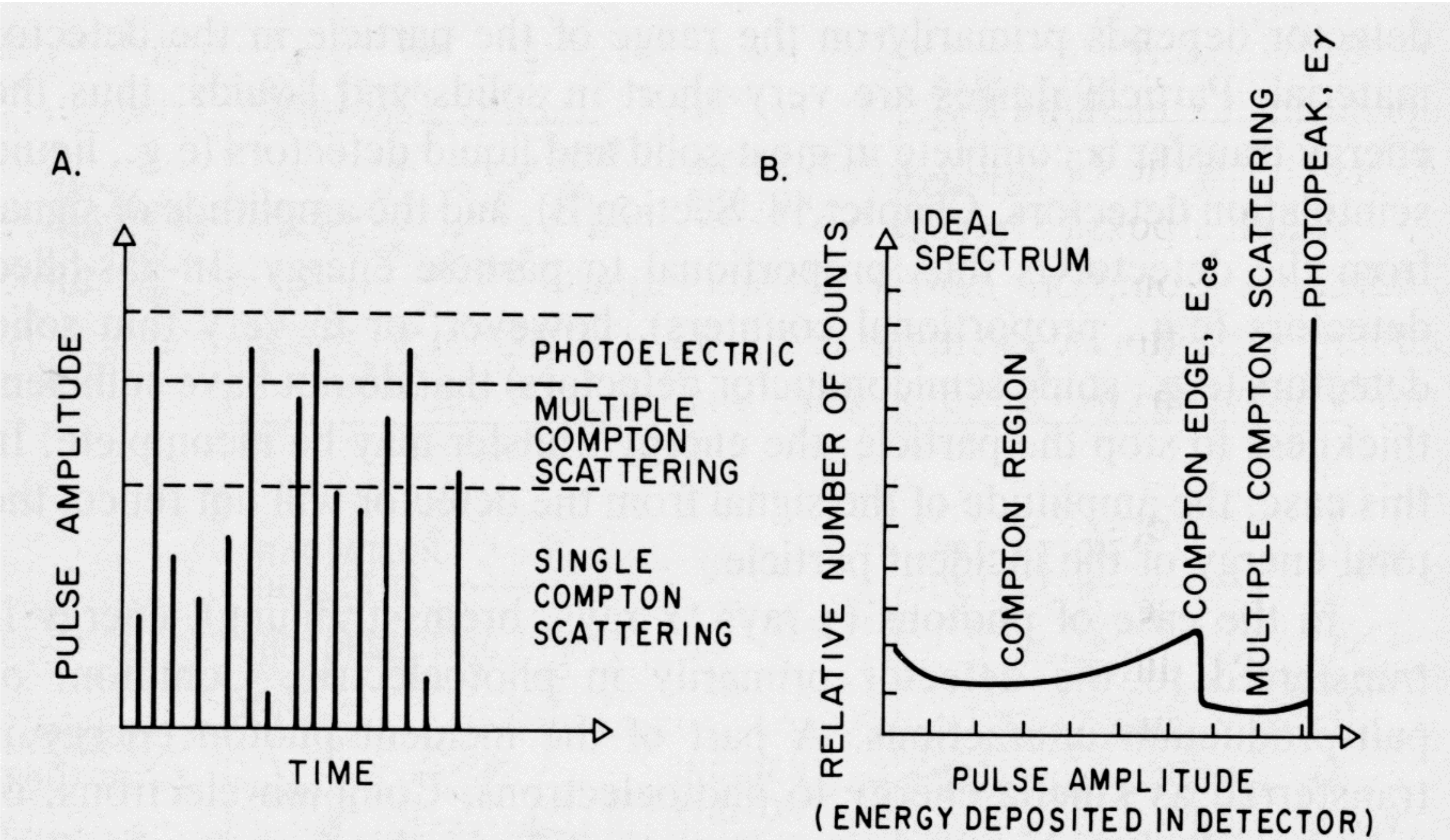

From: Physics in Nuclear Medicine (Cherry, Sorenson and Phelps)

# Sample Spectroscopy System




From: The Essential Physics of Medical Imaging (Bushberg, et al)

# Multichannel Analyzer




From: Physics in Nuclear Medicine (Sorenson and Phelps)

# Energy Resolution

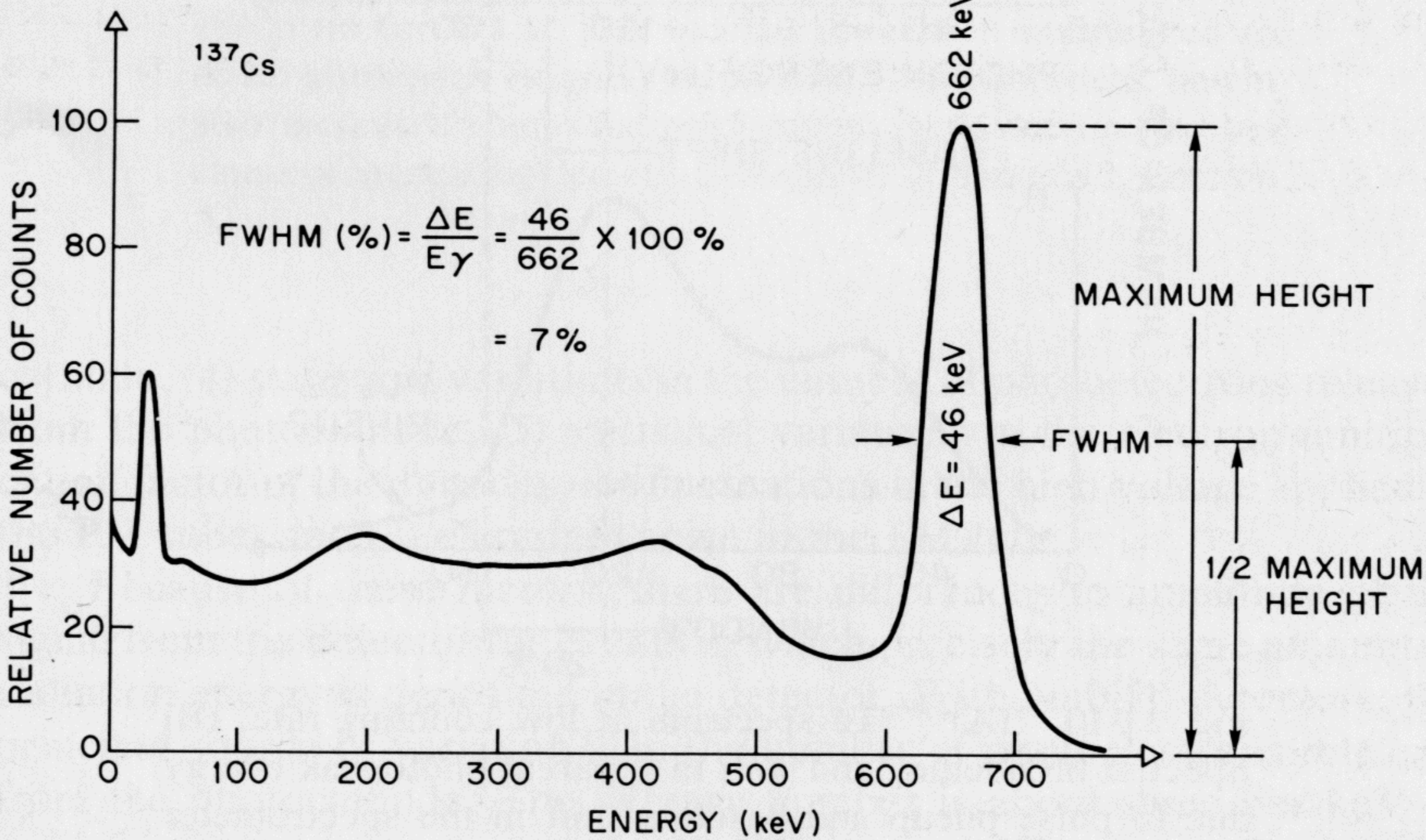



Fig. 11-11. Calculation of FWHM energy resolution of a NaI(Tl) detector for $^{137}Cs$ 662 keV $\gamma$ rays.

From: Physics in Nuclear Medicine (Sorenson and Phelps)

# Question

The count rate for a 1 $\mu$Ci source is measured as 25 kcps by a well counter. Assuming no corrections are applied, the measured count rate for a 10 $\mu$Ci source will be:

a. 250 kcps
b. Less than 250 kcps
c. Greater than 250 kcps

# Questions

A pulse height analyzer (PHA) window can be used to:

a. Identify the energy of a radionuclide
b. Reject Compton scattered photons
c. Separate a mixture of radionuclides
d. Alter the sensitivity or resolution of the system
e. All of the above

From: Raphex

# The Scintillation Camera: Planar

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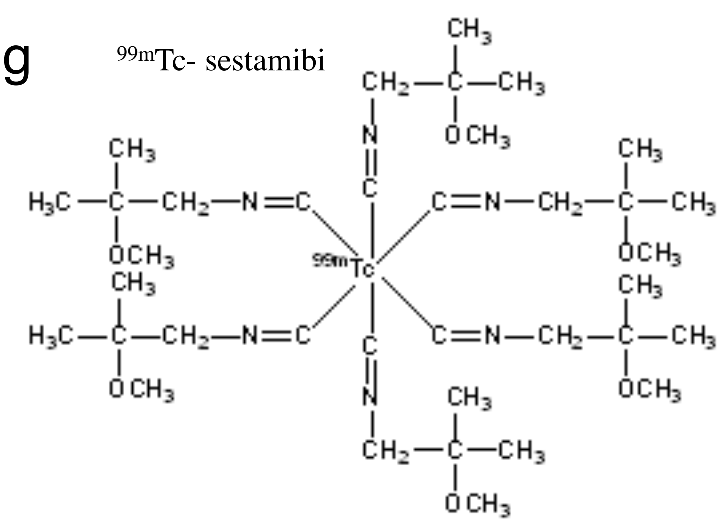
# Radiotracers

- Suitable radionuclides are selected based on
  - photon energy (i.e. ~140 to 511keV)
  - half-life (a few hours)
  - 'clean' decay, i.e. no alpha and beta particles, which only add dose
- The radiotracer (ligand+nuclide) must have suitable biodistribution, clearance, and be safe in 'trace' amounts
- Examples
  - $^{99m}$Tc-labelled Methylene diphosphonate ($^{99m}$Tc-MDP) for bone, where MDP is a phosphate that exchanges with bone growth
  - $^{99m}$Tc-labelled sestamibi for myocardial (cardiac muscle) blood perfusion imaging

$^{99m}$Tc- sestamibi

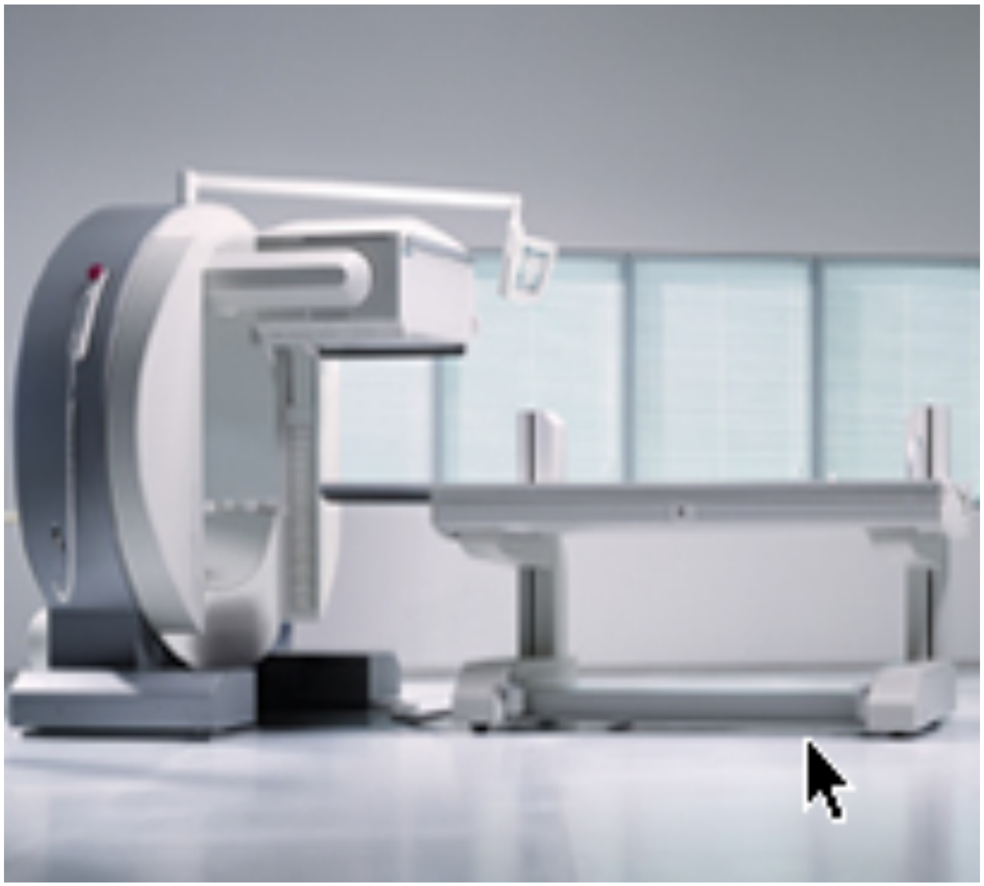


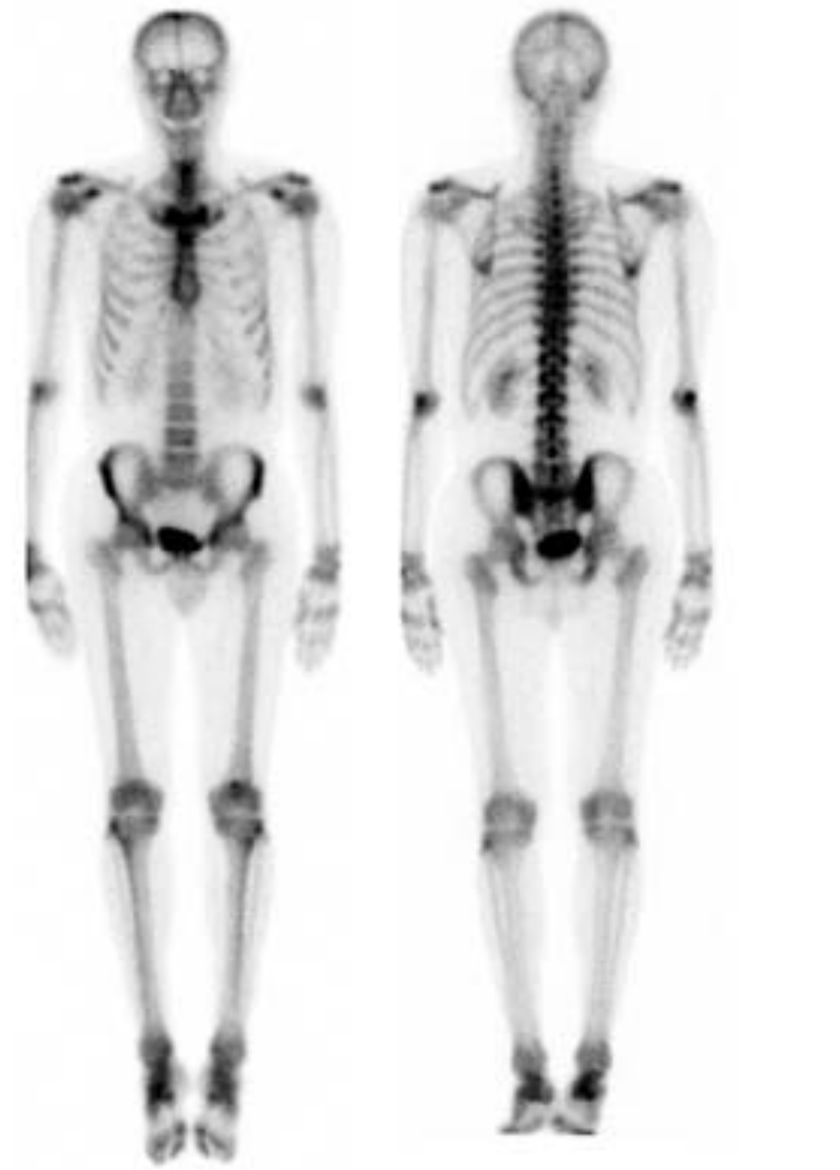
# List of Nuclear Medicine ‘Single Photon’ Radionuclides

| | | |
|---|---|---|
| • Tc99m | 140.5 keV | 6.03 hours |
| • I-131 | 364, 637 keV | 8.06 days |
| • I-123 | 159 keV | 13.0 hours |
| • I-125 | 35 keV | 60.2 days |
| • In-111 | 172, 247 keV | 2.81 days |
| • Th-201 | ~70, 167 keV | 3.044 days |
| • Ga-67 | 93, 185, 300 keV | 3.25 days |

From: Physics in Nuclear Medicine (Sorenson and Phelps)

# The Planar Gamma Camera





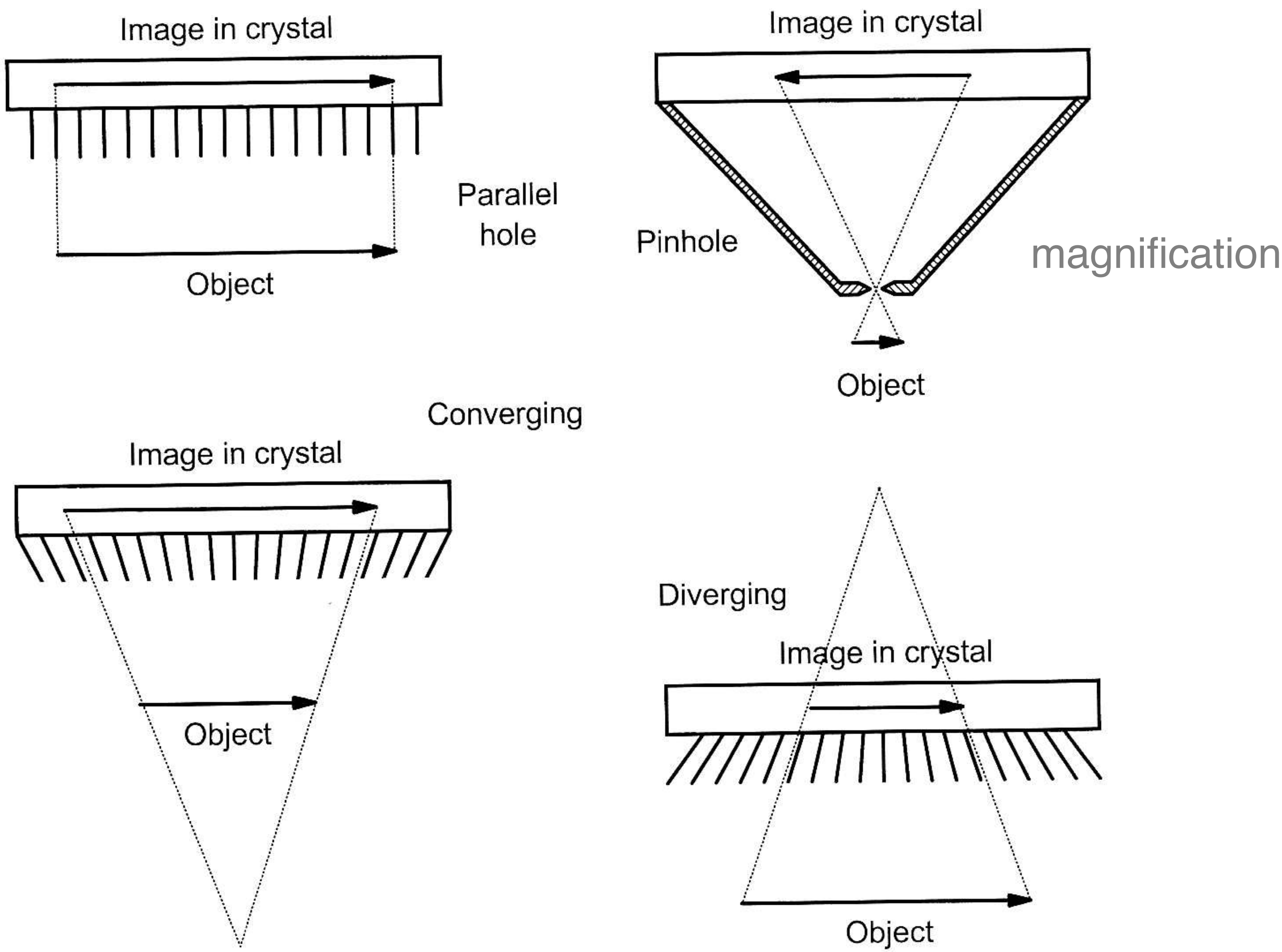
Siemens e.cam

# Instrumentation & Processes

- Detector assembly:
  - Components: Scintillator crystal, Window, Photomultipliers
  - Determines interaction position and energy
  - Performance: Detection efficiency, Intrinsic (spatial) resolution, Energy resolution
- Collimator:
  - Determines Gamma-ray direction
  - Types: Parallel, Converging, Diverging, Pinhole, Multi-pinhole
  - Performance: Penetration, Image Resolution, Efficiency
- Acquisition modes:
  - Frame vs. List mode
  - Static (single), Dynamic (TAC), Gated (cardiac / respiration)
- Camera QA corrections:
  - Uniformity, Linearity, Photo-peak window, Multi-energy registration

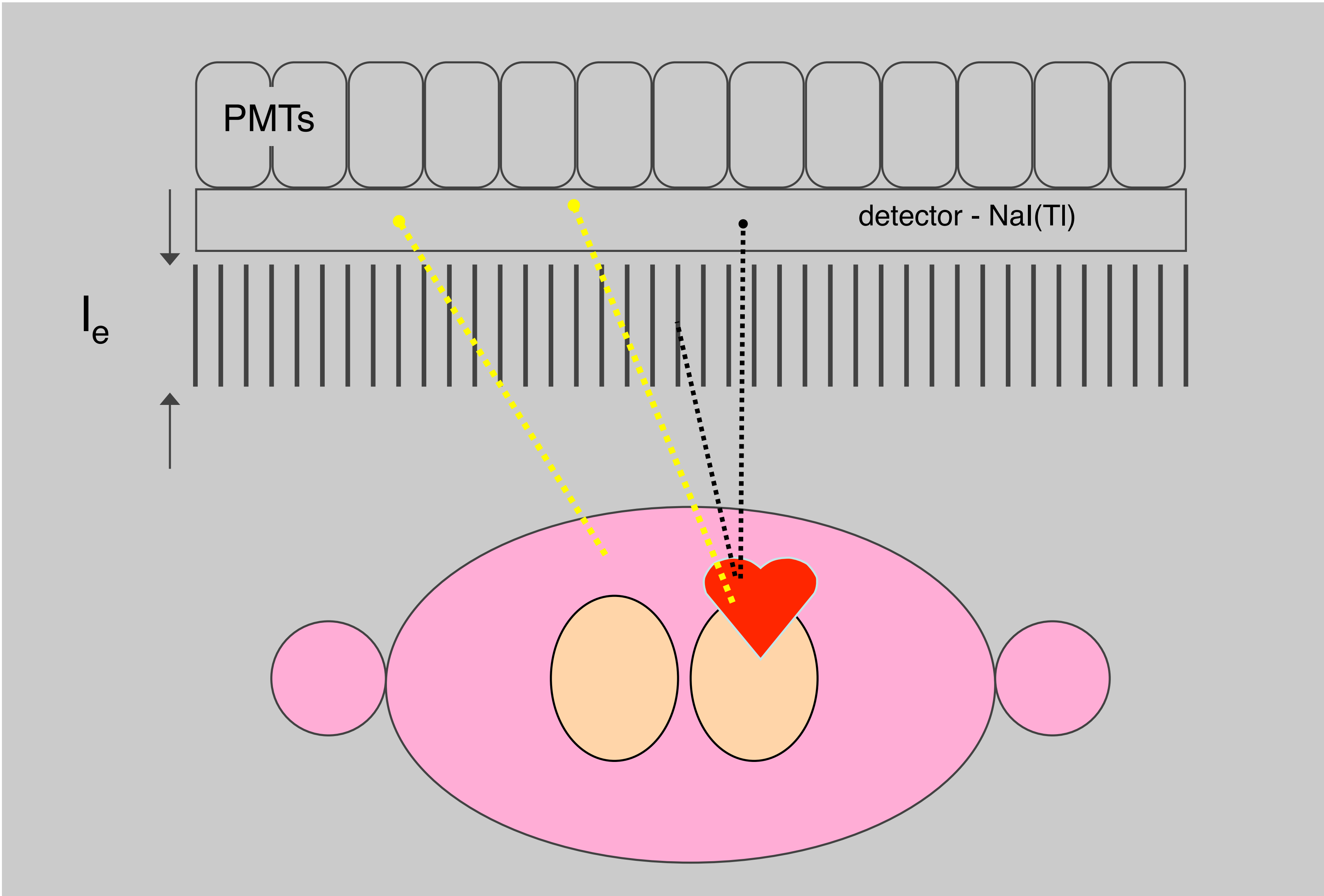
# The Scintillation Camera: Collimators

# Types of Collimators

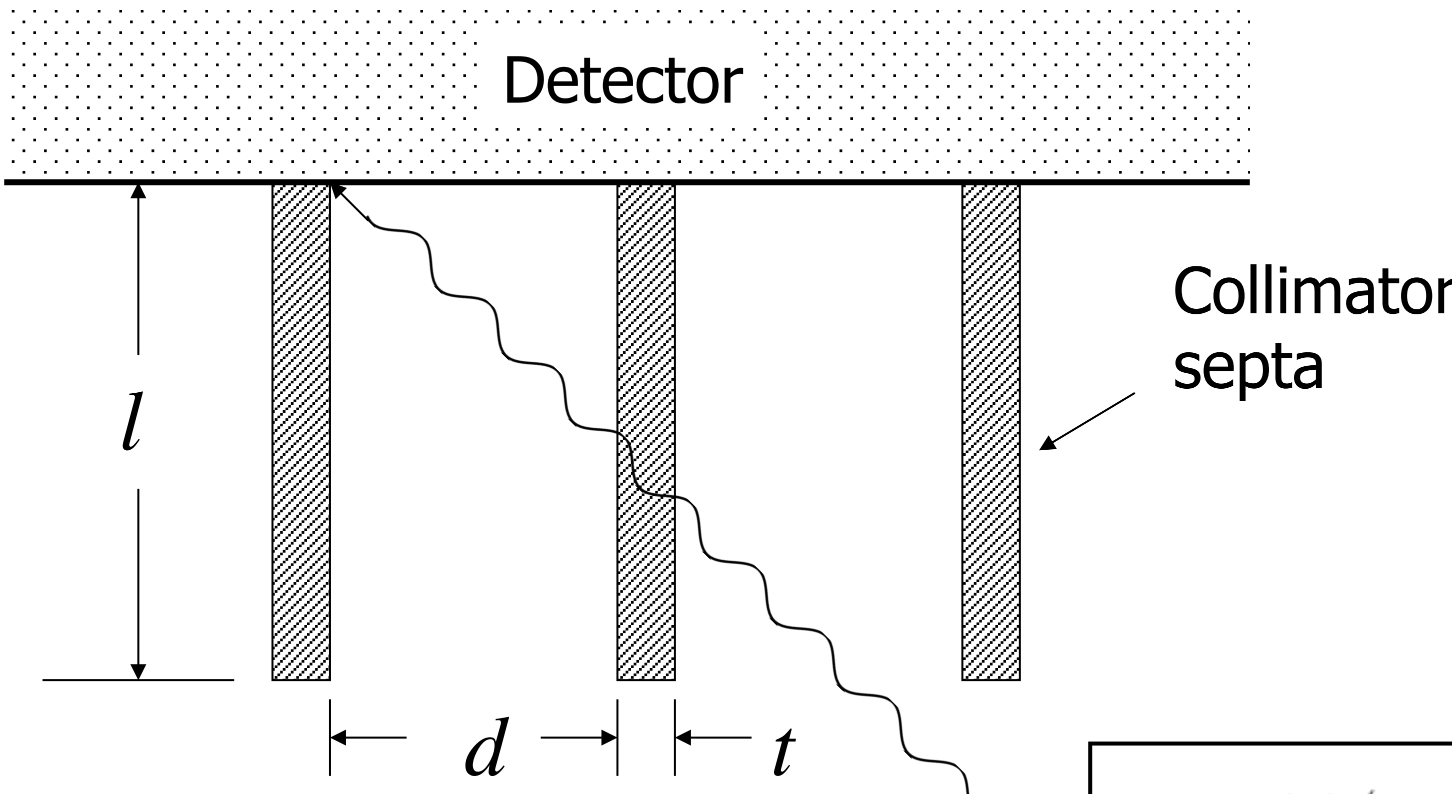



From: The Essential Physics of Medical Imaging (Bushberg, et al)

# Parallel Hole Collimator



# Collimators - Septal Penetration




Minimum septa thickness, t, for <5% septal penetration:

$$t \geq \frac{6d/\mu}{l - \left(3/\mu\right)}$$

From: Physics in Nuclear Medicine (Cherry, Sorenson and Phelps)

# Collimator Efficiency

Collimators typically absorb well over 99.95% of all photons emitted from the patient.

Trade-off between spatial resolution and detection efficiency.

# Collimator Resolution

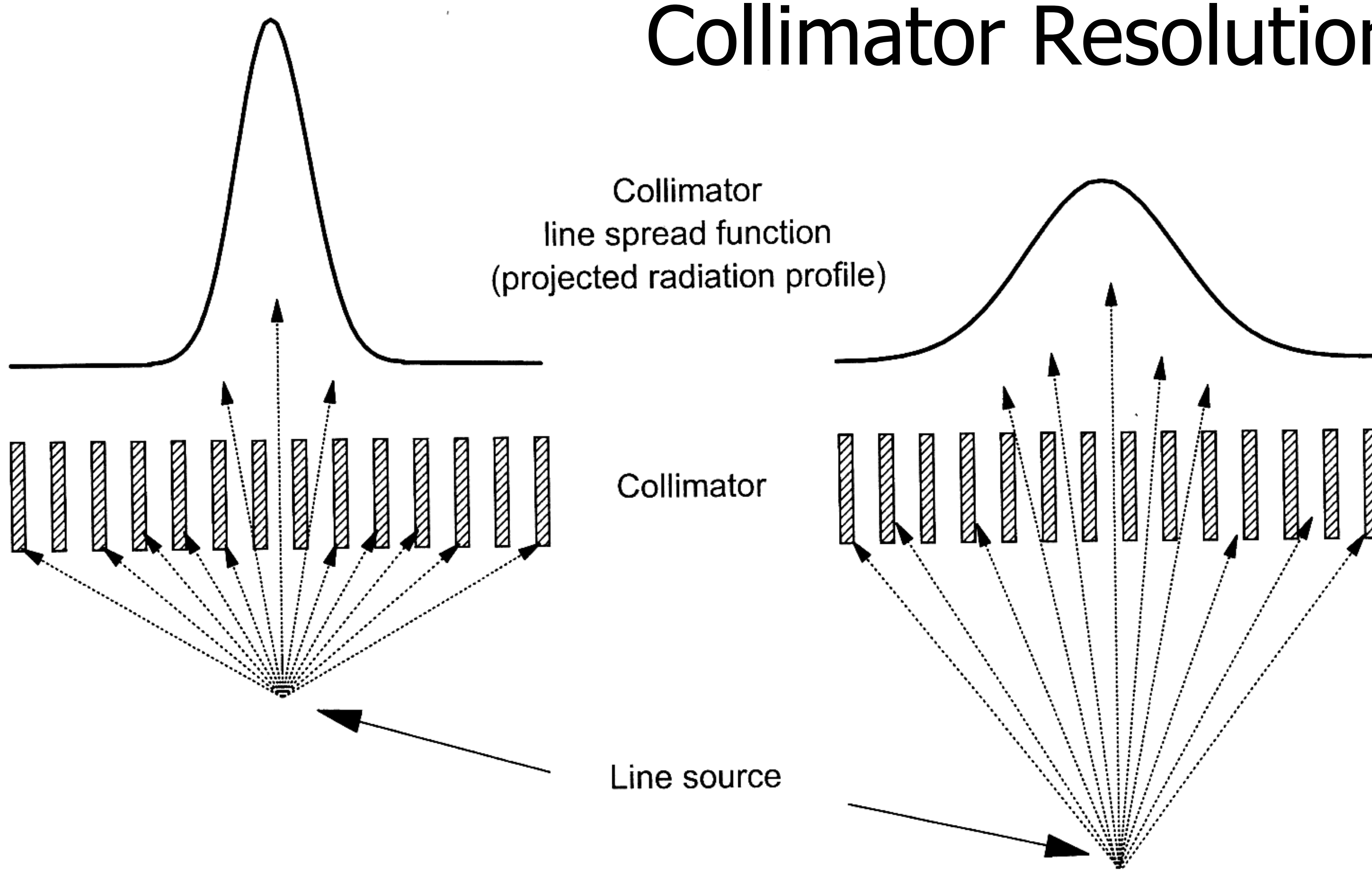
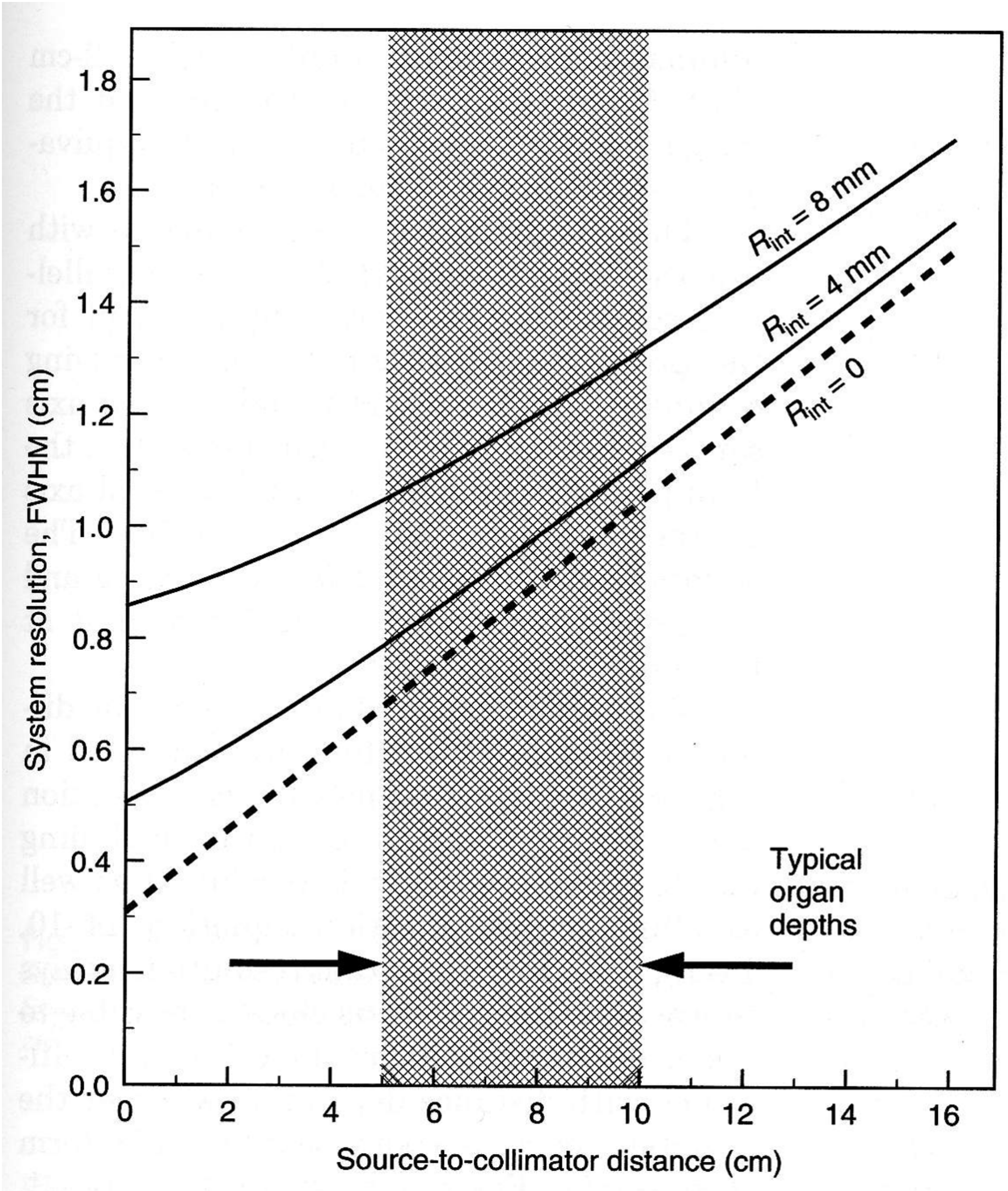



**FIGURE 21-12.** Line spread function (LSF) of a parallel-hole collimator as a function of source-to-collimator distance. The full-width-at-half-maximum (FWHM) of the LSF increases linearly with distance from the source to the collimator; however, the total area under the LSF (photon fluence through the collimator) decreases very little with source to collimator distance. (In both figures, the line source is seen "end-on.")

From: The Essential Physics of Medical Imaging (Bushberg, et al)

# Gamma Camera - spatial resolution




$$R_s = \sqrt{\left(R_i^2 + R_c^2\right)}$$

From: Physics in Nuclear Medicine (Cherry, Sorenson and Phelps)

# Collimator: Resolution and Sensitivity

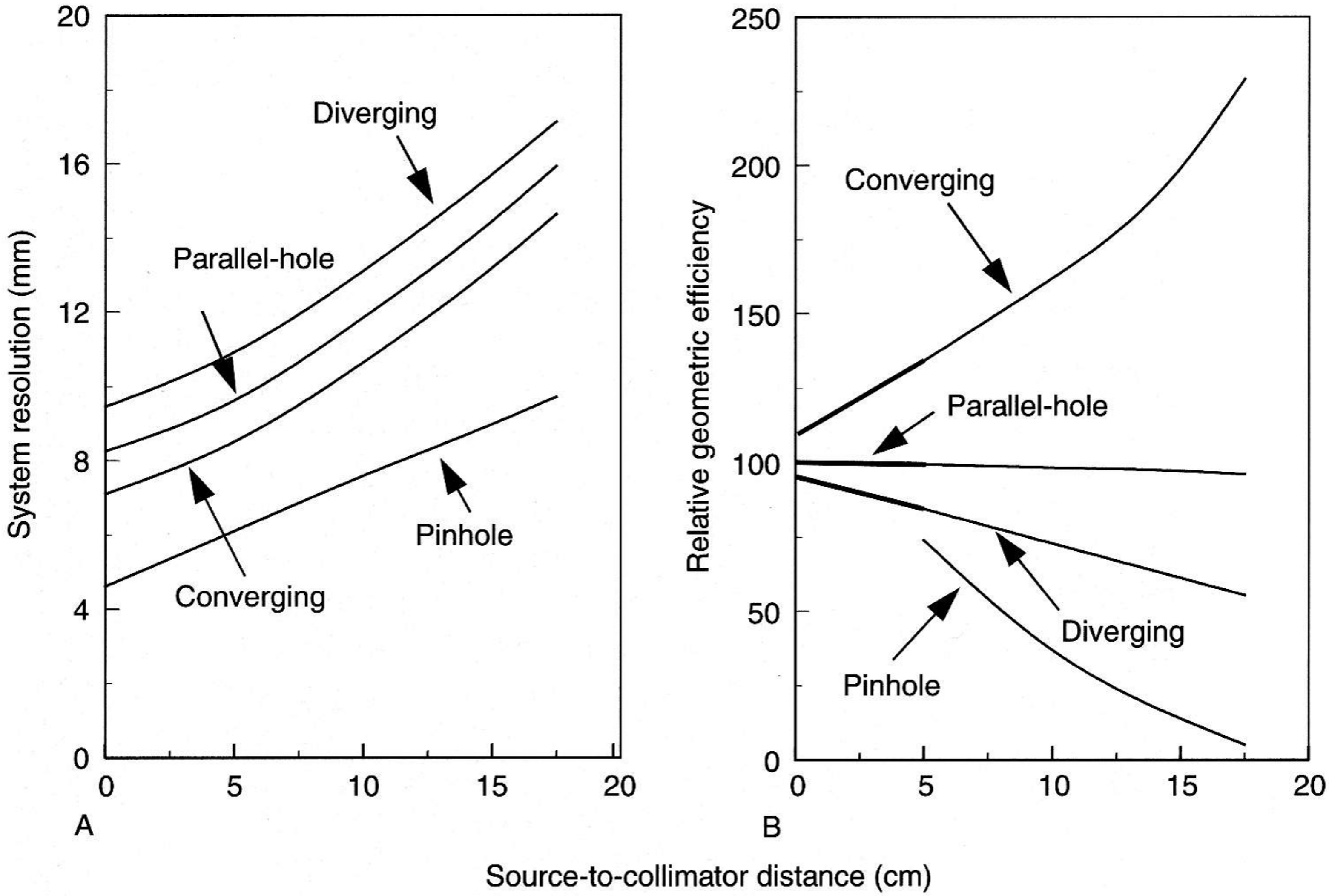



**Figure 14–21.** Performance characteristics (*A*, system resolution; *B*, point-source geometric efficiency in air) versus source-to-collimator distance for four different types of gamma camera collimators. (Reprinted by permission of the Society of Nuclear Medicine from Moyer RA: A low-energy multihole converging collimator compared with a pinhole collimator. J Nucl Med 15:59–64, 1974.)

From: Physics in Nuclear Medicine (Cherry, Sorenson and Phelps)

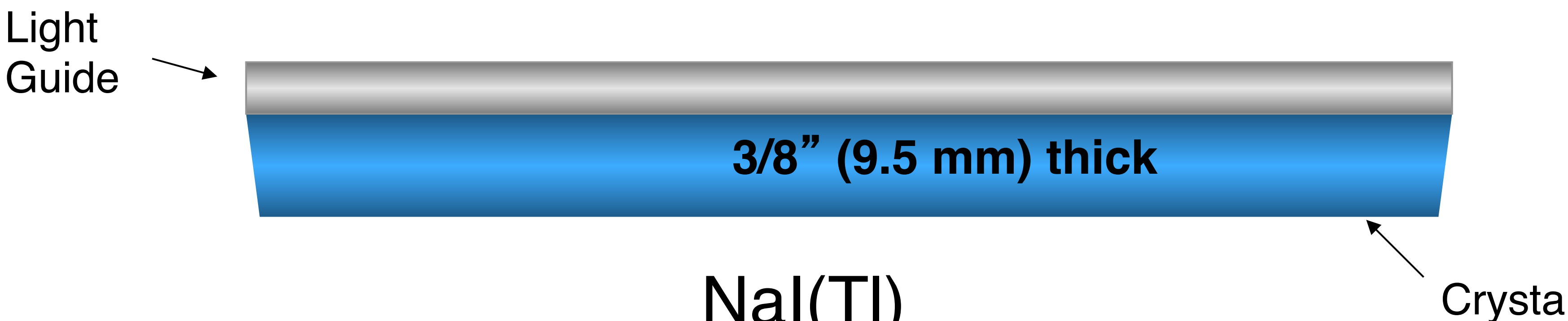
# Collimator choices

| Design | Trade-offs |
|---|---|
| Longer, thicker septa | Reduced penetration, but lower sensitivity |
| Shorter, thinner septa | Increased penetration, but higher sensitivity |
| Longer, thicker septa with wider holes | Reduced penetration, but lower resolution used for high energy photons |
| Shorter, thinner septa with narrower holes | Increased penetration, but higher resolution used for lower energy photons |

| Collimator | Max energy [keV] | Resolution (FWHM at 10 cm) |
|---|---|---|
| LEHR (low-energy, high resolution) | 150 | 7.4 mm |
| LEAP or LEGP (low-energy, general purpose) | 150 | 9.1 mm |
| Medium Energy | 400 | 13.2 mm |
| High Energy/Ultra High Energy (PET Scanning) | 511 | |

# The Scintillation Camera: Detector System

# Crystal and light guide




NaI(Tl)

| | |
|---|---|
| Density | 3.67 g/cm$^3$ |
| Attenuation Coefficient (@140 keV) | 2.64 cm$^{-1}$ |
| PE fraction | ~80% |
| Light output | 40K/MeV |
| Decay time | 230 nsec |
| Wavelength | 410 nm |

# Detection Efficiency

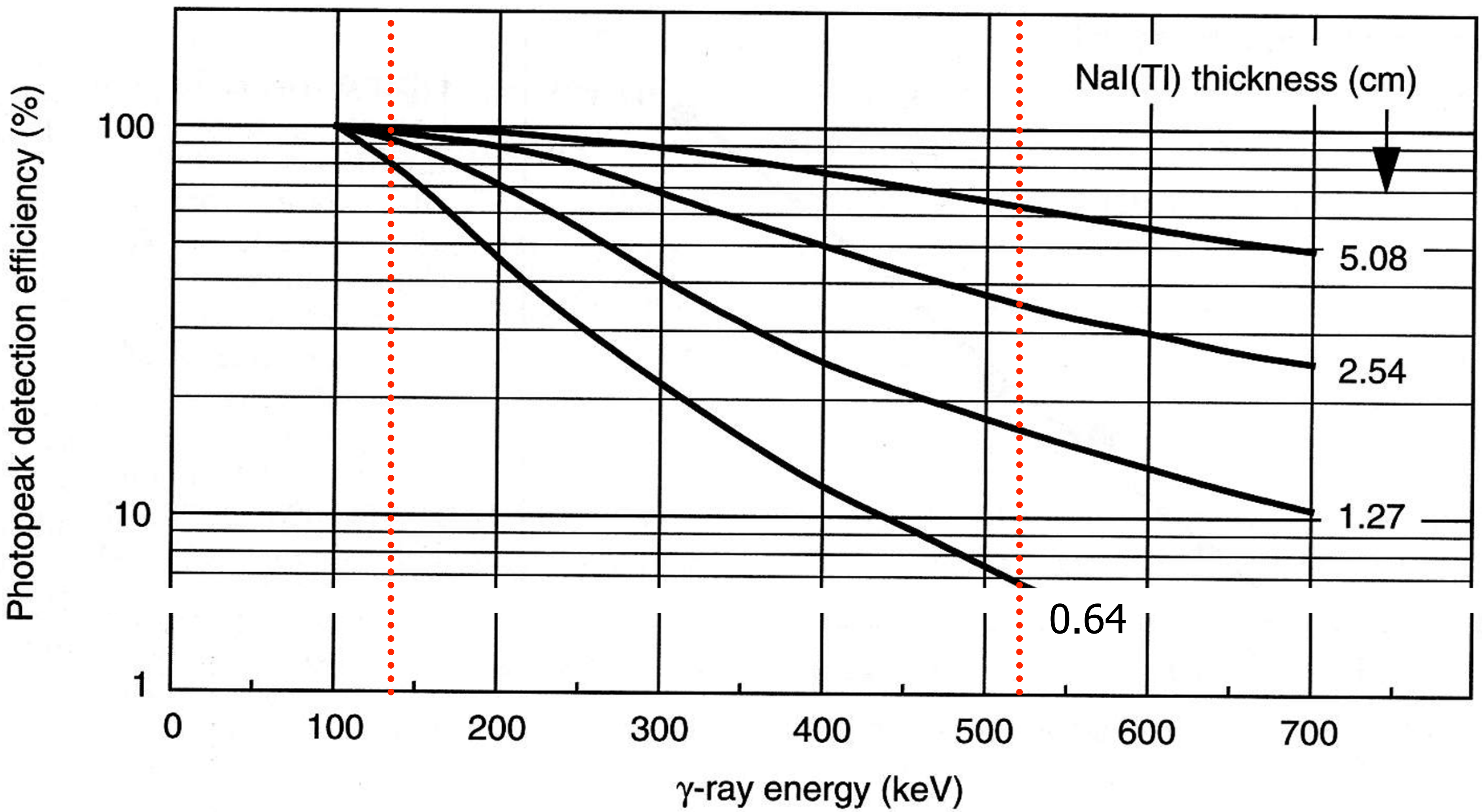



**Figure 14–3.** Photopeak detection efficiency versus γ-ray energy for NaI(Tl) detectors of different thicknesses. (Adapted from Anger HO: Radioisotope cameras. In Hine GJ [ed]: Instrumentation in Nuclear Medicine, Vol 1. New York, Academic Press, 1967, p 506.)

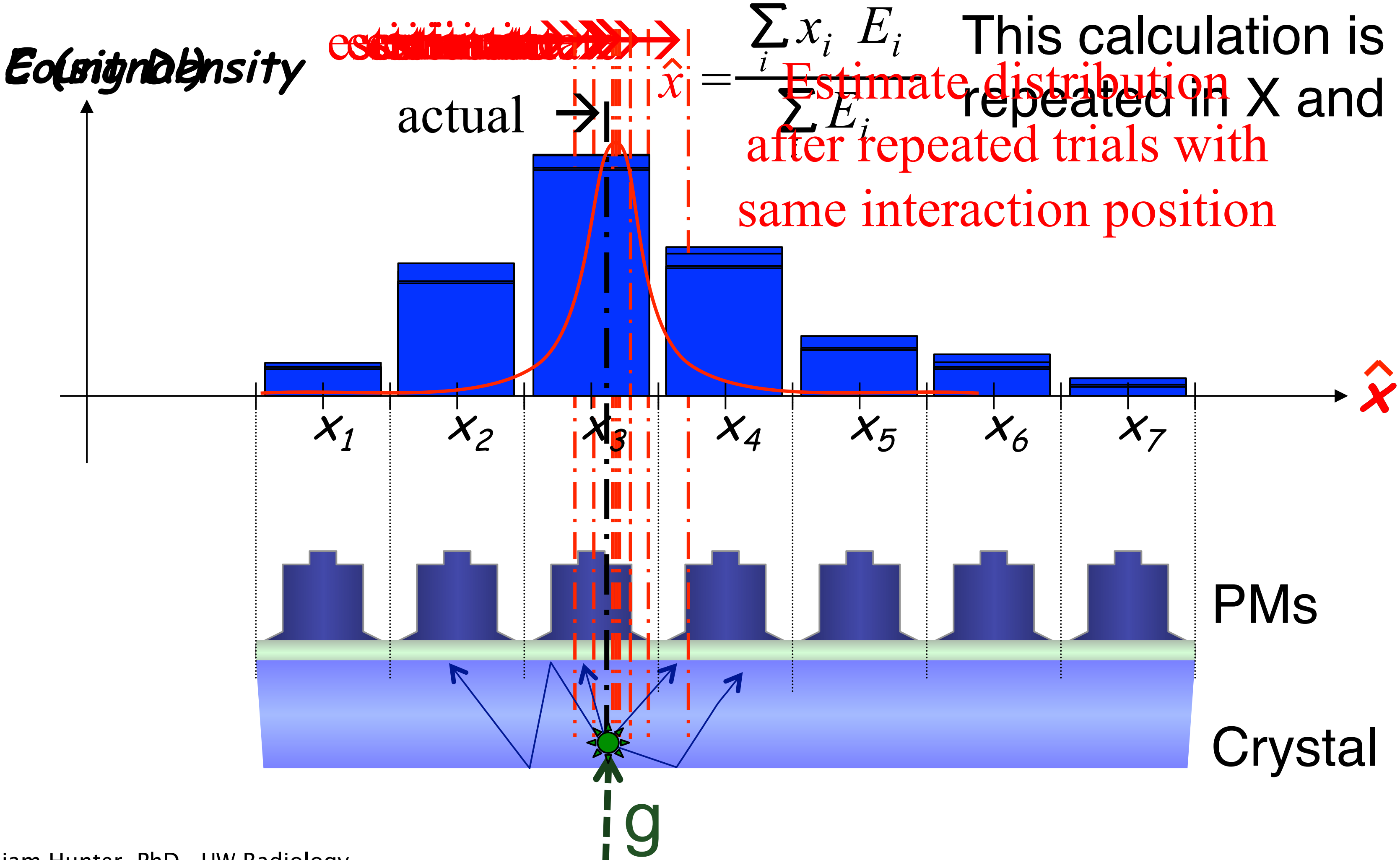
# Event Positioning in Detectors

- Detect gamma rays & determine position using light distribution
- For example, we could use the energy centroid
- Intrinsic resolution <4 mm FWHM for current clinical systems
- Signal noise will blur event positioning.

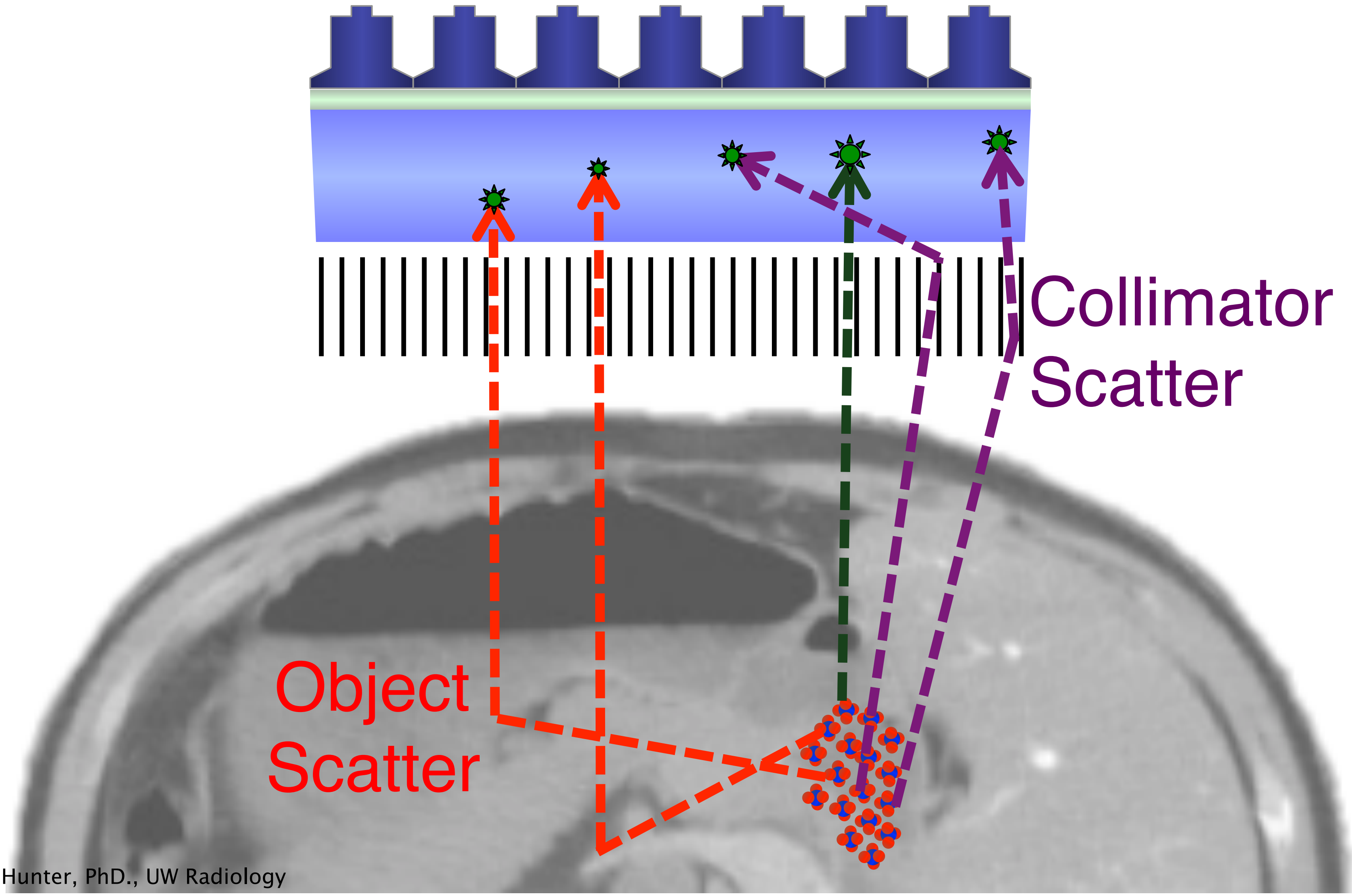
$$\hat{x} = \frac{\sum_i x_i E_i}{\sum_i E_i}$$

This calculation is repeated in X and Y

Estimate distribution after repeated trials with same interaction position

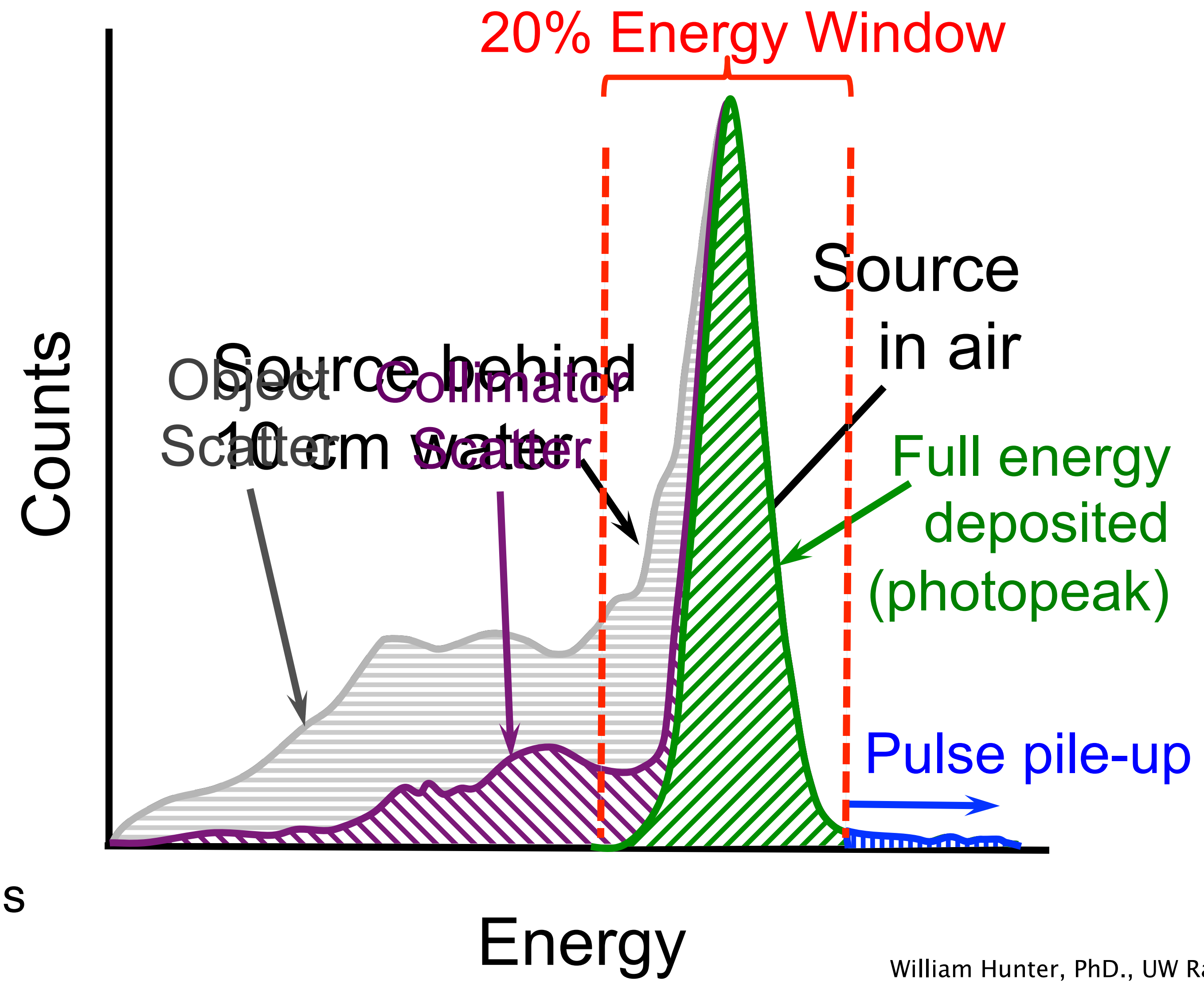


# Object & Collimator Scatter




William Hunter, PhD., UW Radiology

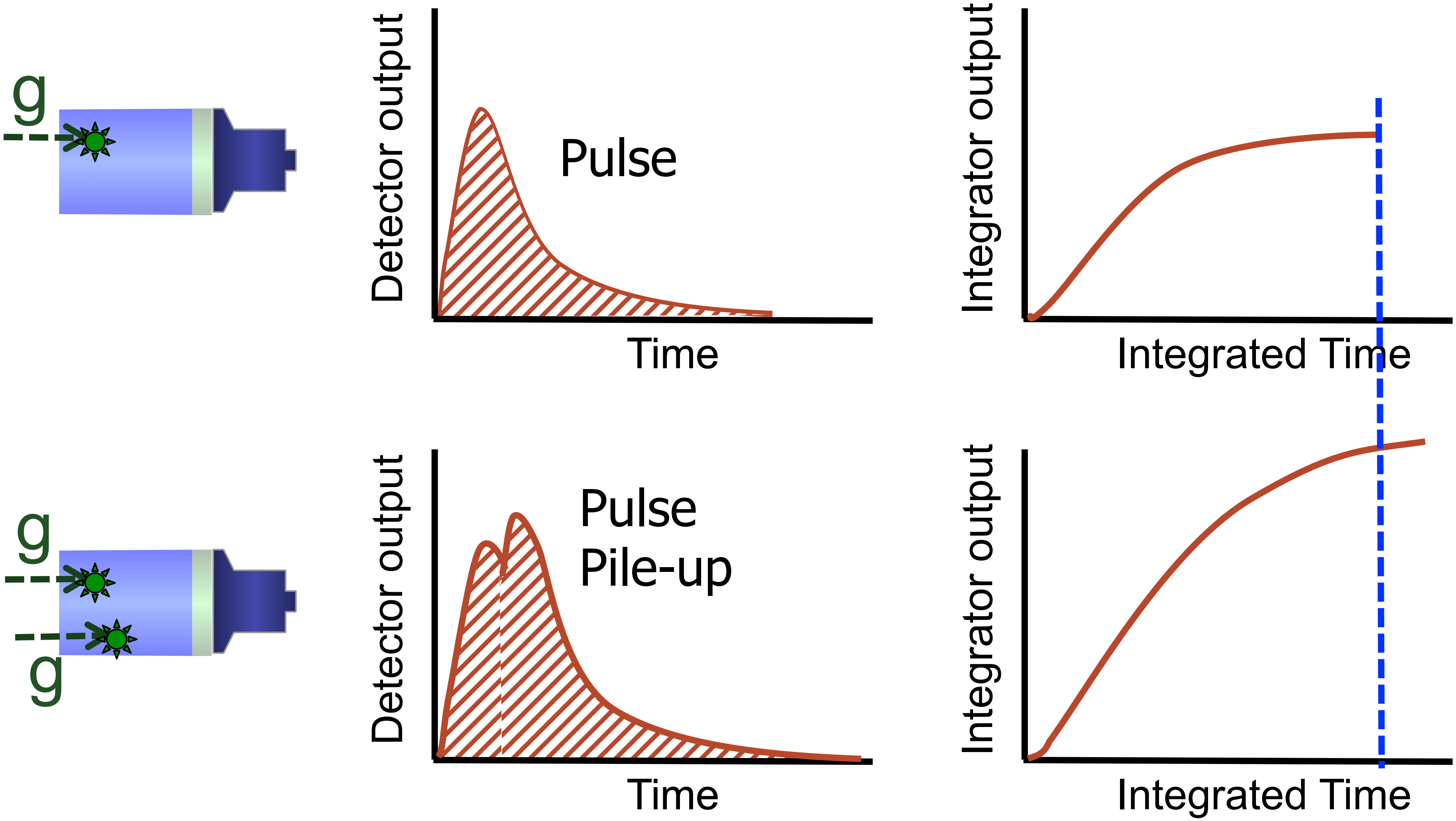
# Gamma Camera Energy Spectra




140 keV photons
9.5 mm crystal

William Hunter, PhD., UW Radiology

# Pulse pile-up




- Occurs more often at higher count rates and longer integration time
- Extends "dead time", paralyzing the detector & decreasing sensitivity

# Pulse Pile-up

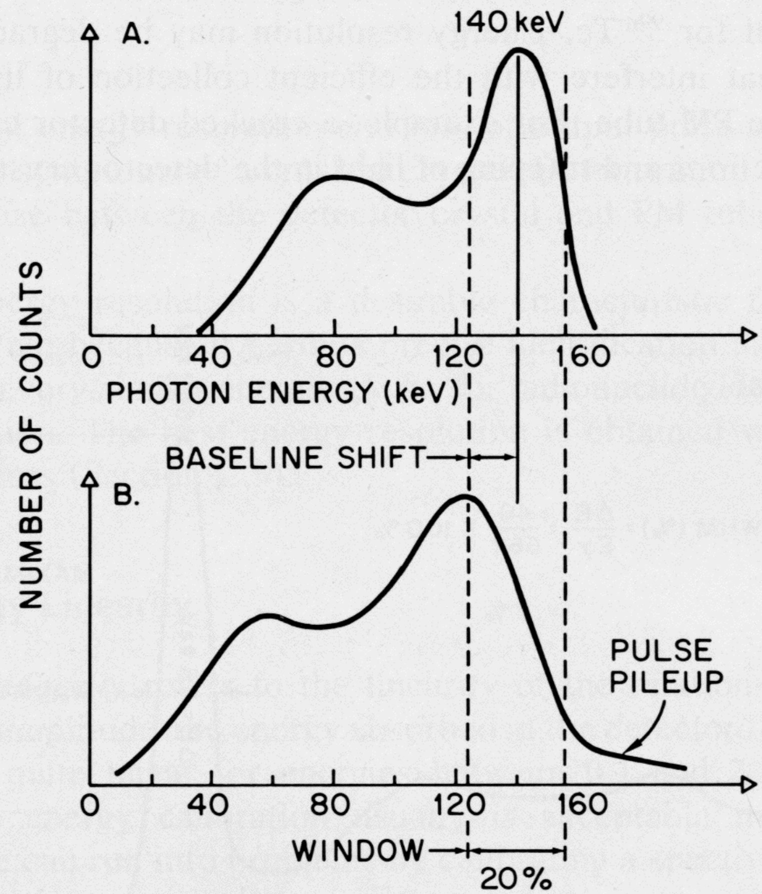



Fig. 11-10. (A) $^{99m}$Tc spectrum at low counting rate. (B) Spectral broadening and shift in apparent photopeak energy due to pulse pileup and baseline shift in the spectrometer amplifier at high counting rate.

## Energy spectra

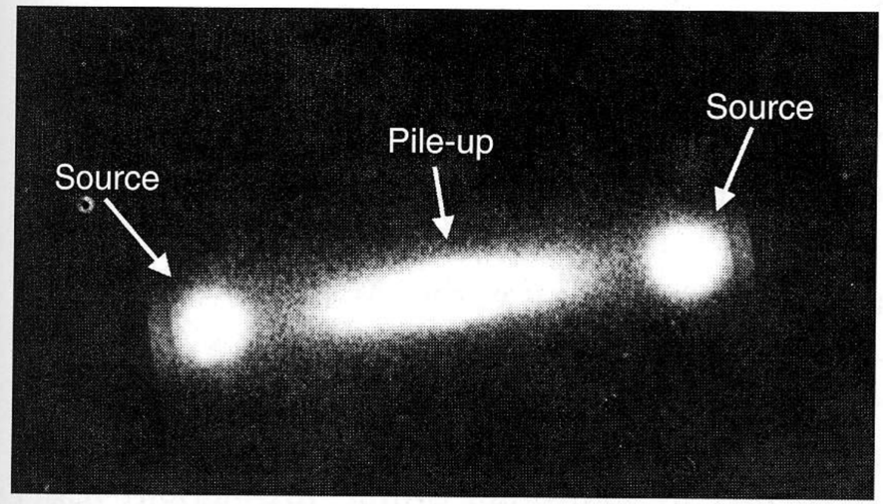
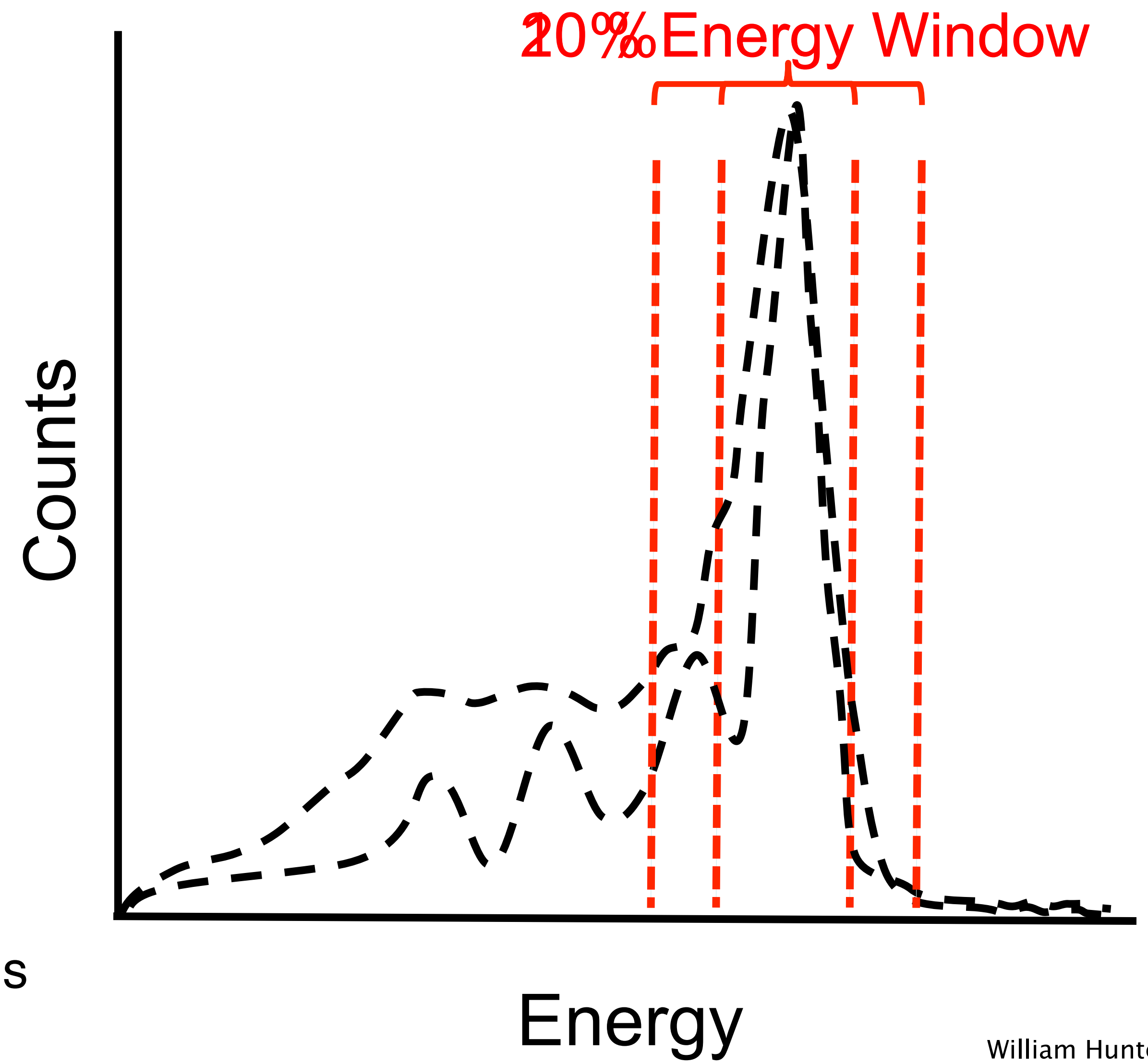



**Figure 14–6.** Images of two $^{99m}$Tc point sources of relatively high activities (~370 MBq each). Events appearing in the band between the two point-source locations are mispositioned events due to pulse pile-up.

## Pile-up in image

From: Physics in Nuclear Medicine (Sorenson and Phelps) and (Cherry, Sorenson and Phelps)

# Improved scatter rejection with better energy resolution




140 keV photons
9.5 mm crystal

William Hunter, PhD., UW Radiology

# Energy Windows

Most gamma cameras can acquire data using multiple energy windows.  Allows for simultaneous imaging of different radioisotopes, for example Tc-99m (140 keV) and F-18 (511 keV).

# Raphex Question

**D67.** A patient with a history of thyroid cancer has suspected bone marrow metastases in the cervical spine. It is recommended to perform both an I-131 radioiodine scan as well as a bone scan using the Tc-99m-MDP. Which would be the optimum sequence to perform unambiguous scans in the ***shortest*** time?

A. Administer the I-131 and Tc-99m simultaneously. Perform the bone scan first and recall the patient after 24 hours for the radioiodine scan.
B. Administer the I-131 first. Perform the I-131 thyroid scan at 24 hours, then inject Tc-99m MDP and perform the bone scan shortly afterwards.
C. Administer the I-131 first. Perform the I-131 thyroid scan at 24 hours, then ask the patient to wait 3 to 6 weeks until the I-131 has fully decayed before performing the bone scan.
D. Administer the Tc-99m MDP first. Perform the bone scan. Then administer the I-131, and perform the thyroid scan after 24 hours.
E. Administer the Tc-99m MDP first, followed shortly thereafter by the I-131. Then perform the bone scan followed by the thyroid scan after 24 hours.

# Raphex Answer

**D67. D**

The presence of I-131 will interfere with a Tc-99m bone scan but not vice versa. This is because the higher energy 364 keV I-131 photons down-scatter into the Tc-99m window, while the reverse is not physically possible. Therefore, the Tc-99m must be administered and scanned first. Answer C would work, but would not optimize the time.

# Raphex Question

**D75.** In an anterior spot image of the thyroid, a starburst artifact may be seen. The cause of this artifact is:

A. Contamination of the collimator.
B. Imperfections in the evenness of the collimator holes.
C. An image reconstruction artifact caused by filtered back projection.
D. Local photomultiplier tube dead time.
E. Septal penetration.

# Raphex Answer

**D75. E**

Septal penetration occurs when photons travel the shortest distance through the lead collimator, i.e., jump between adjacent collimator holes. The star-like appearance is caused by the hexagonal arrangement of holes in the collimator. A and B would not cause star-shaped artifacts. C gives star-shaped artifacts in PET and SPECT, but an anterior spot view does not require reconstruction. Dead-time leads to a loss of sensitivity.

# Raphex Question

**D64.** What would be the appearance of a gamma camera image if a Tc-99m isotope scan were performed for the same duration but with the wrong collimator: a medium-energy general-purpose instead of a low-energy general-purpose collimator ?

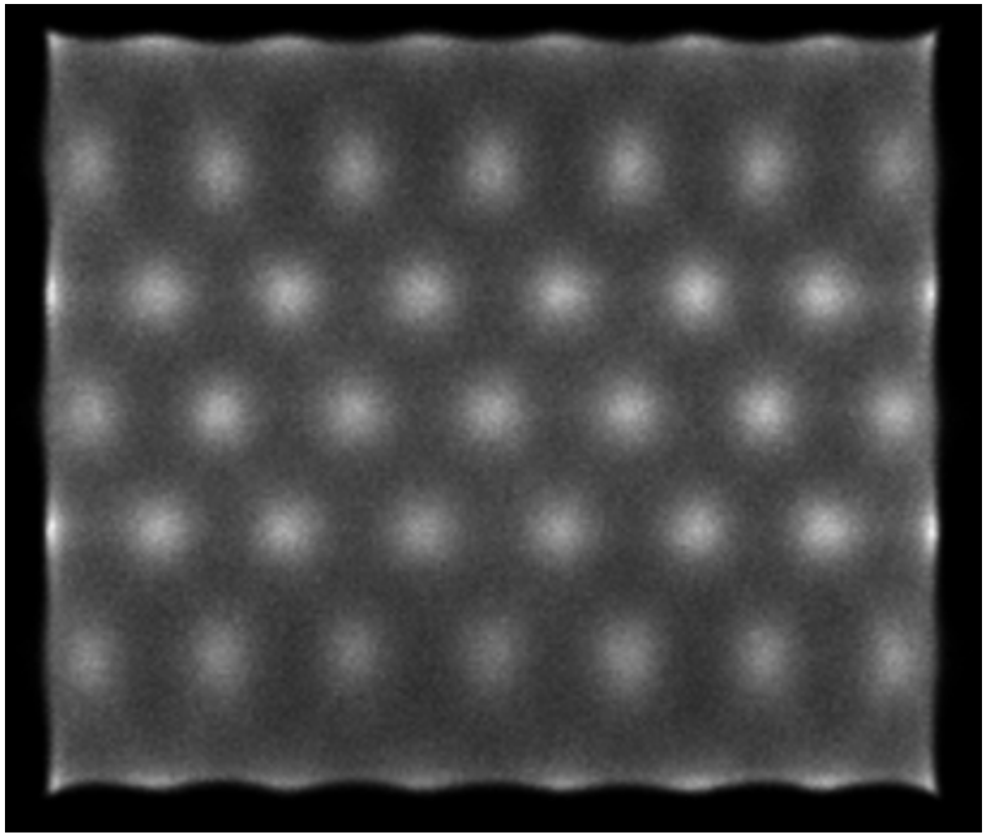
A. There would be absolutely no effect.
B. The image will be more noisy, but probably clinically acceptable.
C. The image quality would be poor due to septal penetration. The study would need to be repeated.
D. There would be so few counts that the study would need to be repeated.
E. This mistake could never happen, because instrument interlocks would prevent a Tc-99m study being performed with the wrong collimator.

# Raphex Answer

**D64. B**

The thicker septa (and smaller hole diameters) of a medium-energy collimator would diminish the count rate by approximately 30% and render the image more statistically noisy. This is less serious than i f a low-energy collimator were used for a medium-energy isotope. In this case, significant septal penetration would substantially degrade the image contrast and render the image unreadable. Whereas the selection of the wrong collimator for a specified isotope would give the technologist a warning message, it would not p revent the gamma camera from acquiring an image in this configuration.

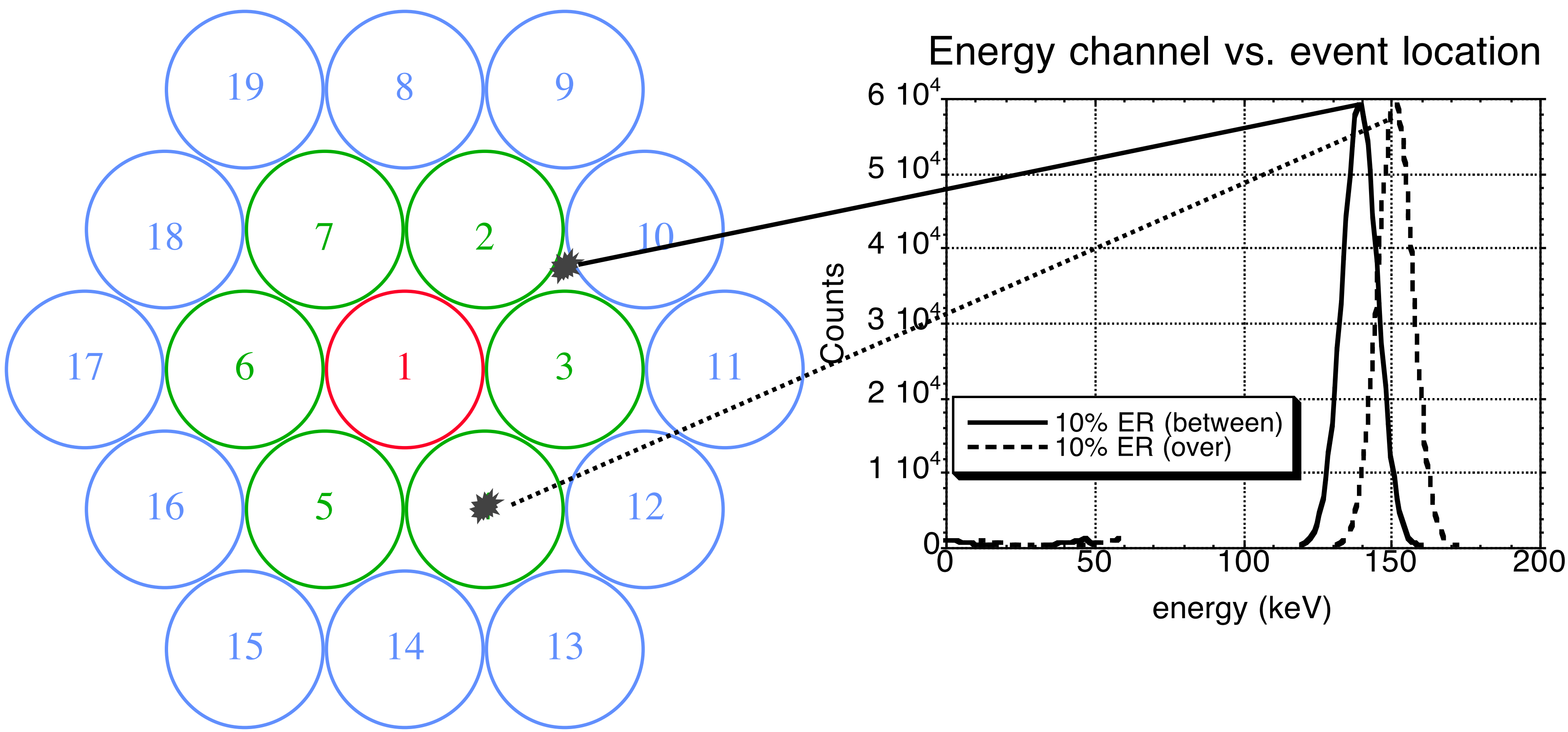
# The Scintillation Camera: Corrections and QA



How was this image produced?

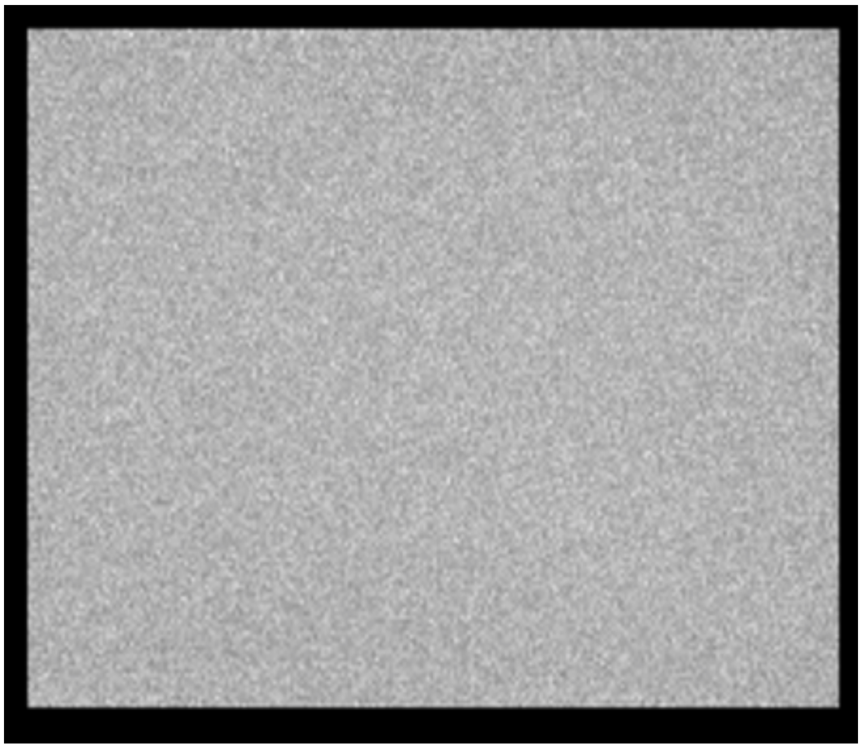
# Gamma Camera Processing Electronics

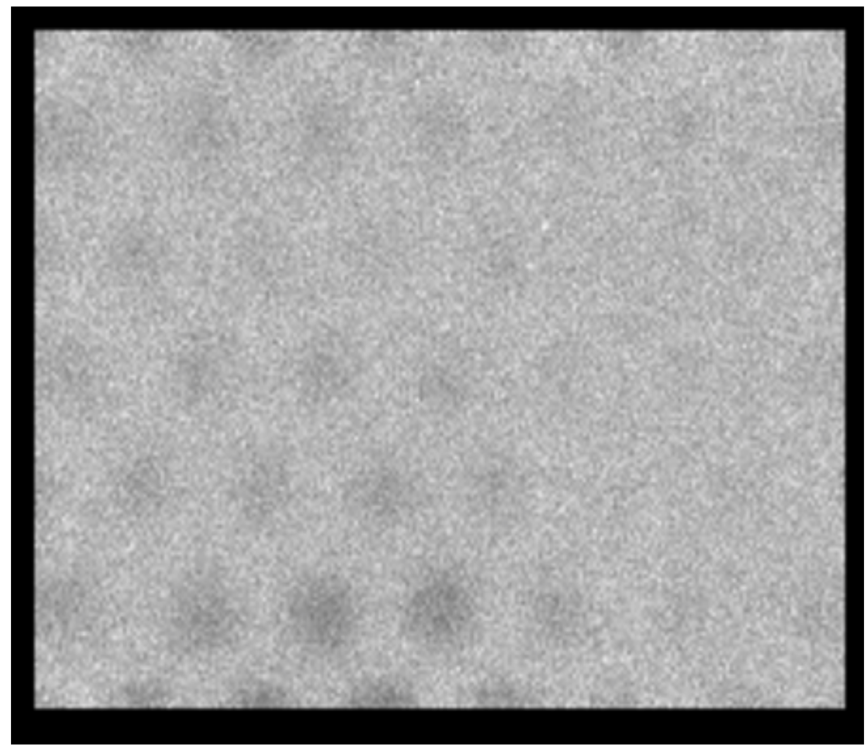
(energy correction)



# Gamma Camera Processing Electronics

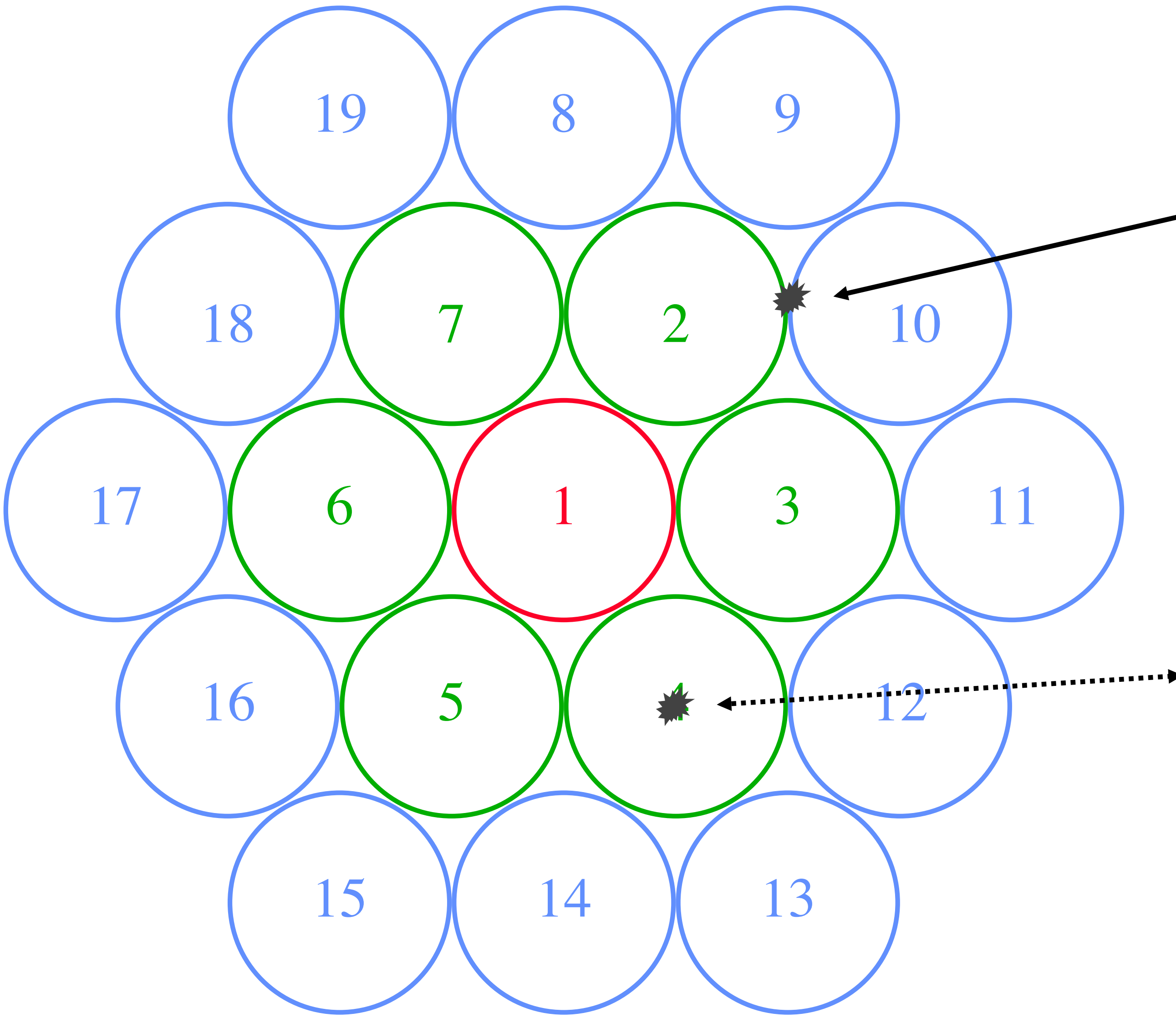
(with and without energy correction)

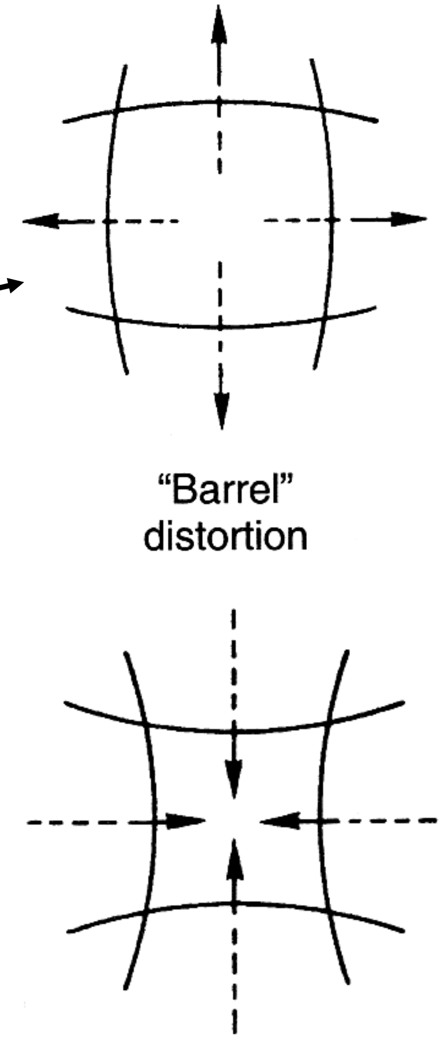




# Gamma Camera Processing Electronics

(linearity correction)



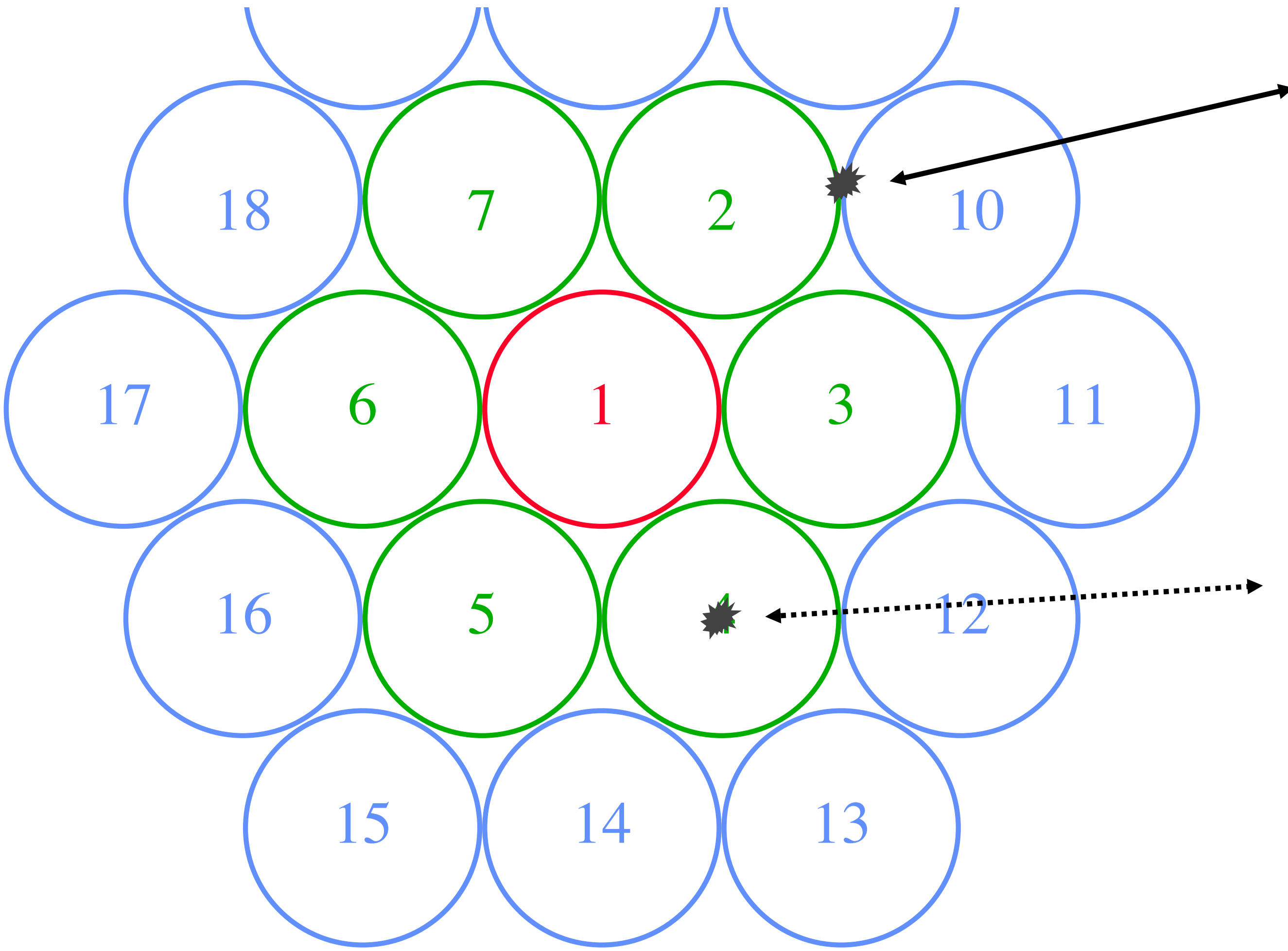



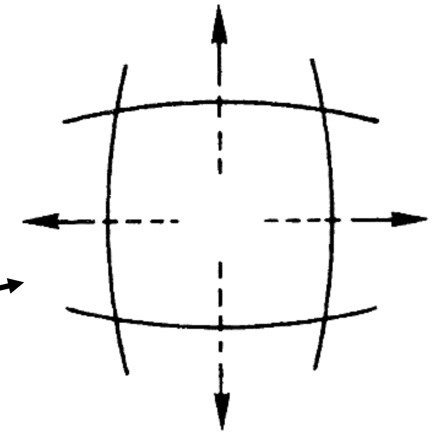

From: Physics in Nuclear Medicine (Cherry, Sorenson and Phelps)

# Gamma Camera Processing Electronics

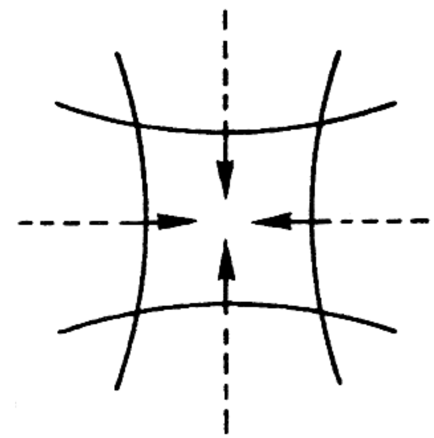
(linearity correction)

Where is the spatial resolution better, directly over a PMT or between PMTs?



"Barrel" distortion

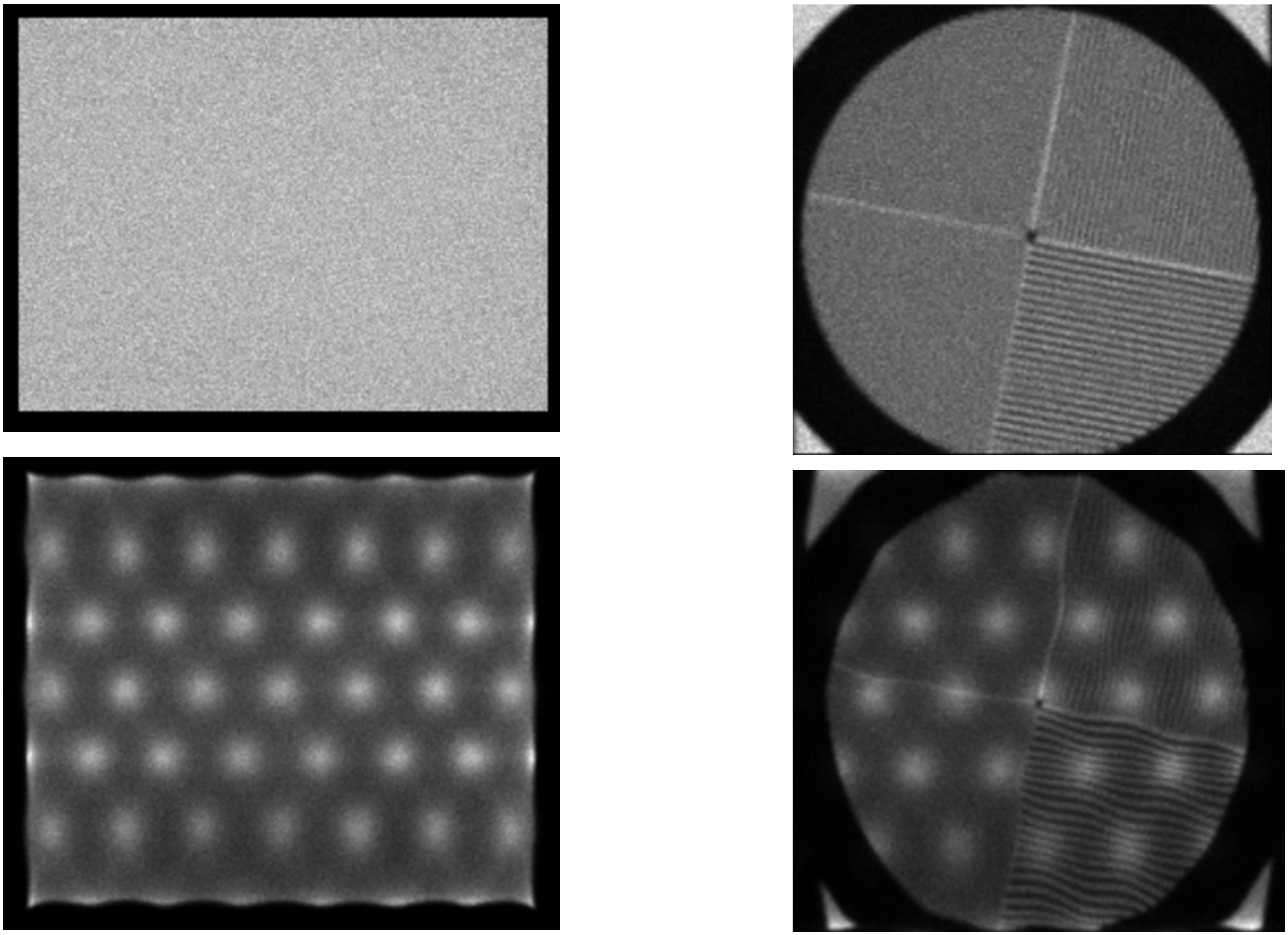


"Pincushion" distortion

From: Physics in Nuclear Medicine (Cherry, Sorenson and Phelps)

# Gamma Camera Processing Electronics

(linearity correction)



# Additional Gamma Camera Corrections
## (sensitivity / uniformity)

Acquired from long uniform flood after energy and linearity corrections have been applied

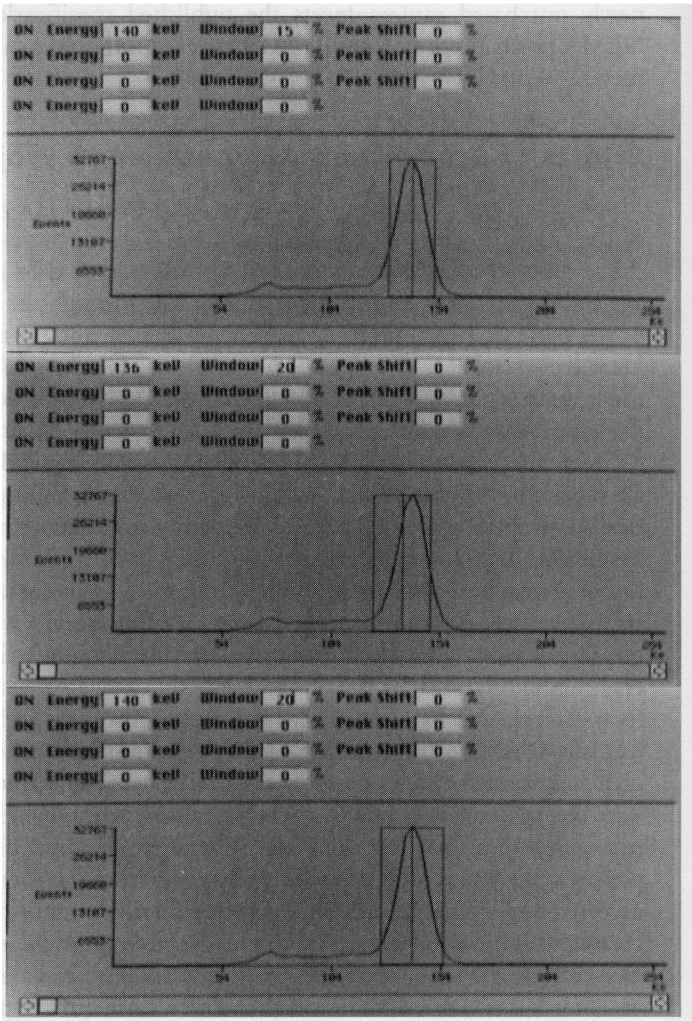
Multiplicative correction

Adjusts for slight variation in the detection efficiency of the crystal

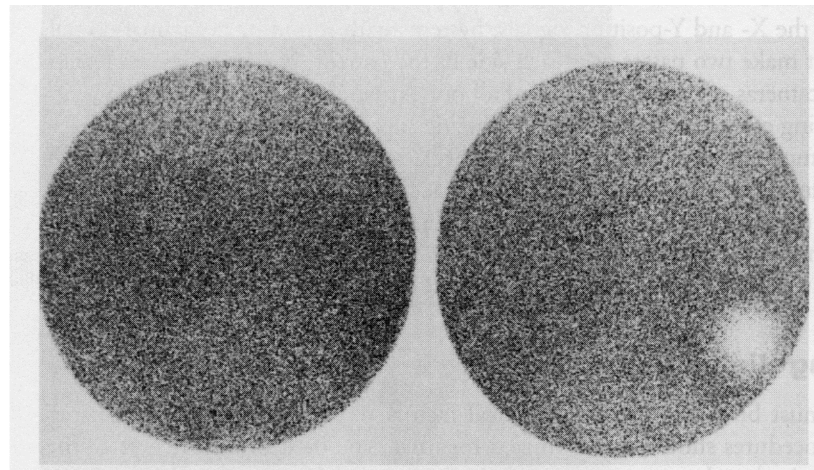
Compensates for small defects or damage to the collimator

Should not be used to correct for large irregularities

# Daily Gamma Camera QA Tests




Photopeak window



Flood uniformity

From: The Essential Physics of Medical Imaging (Bushberg, et al)

# Camera Testing

***Weekly***: *Visual check of the spatial resolution and the geometric linearity*

- Resolution & Linearity: Use Tc-99m flood source, quadrant bar phantom, and LEHR collimator

***Monthly***: *Uniformity of camera for various isotopes & collimators*

- Expected variations in the integral and differential uniformity < 10%.
- Reveals damage to collimators or energy-dependent difference in performance

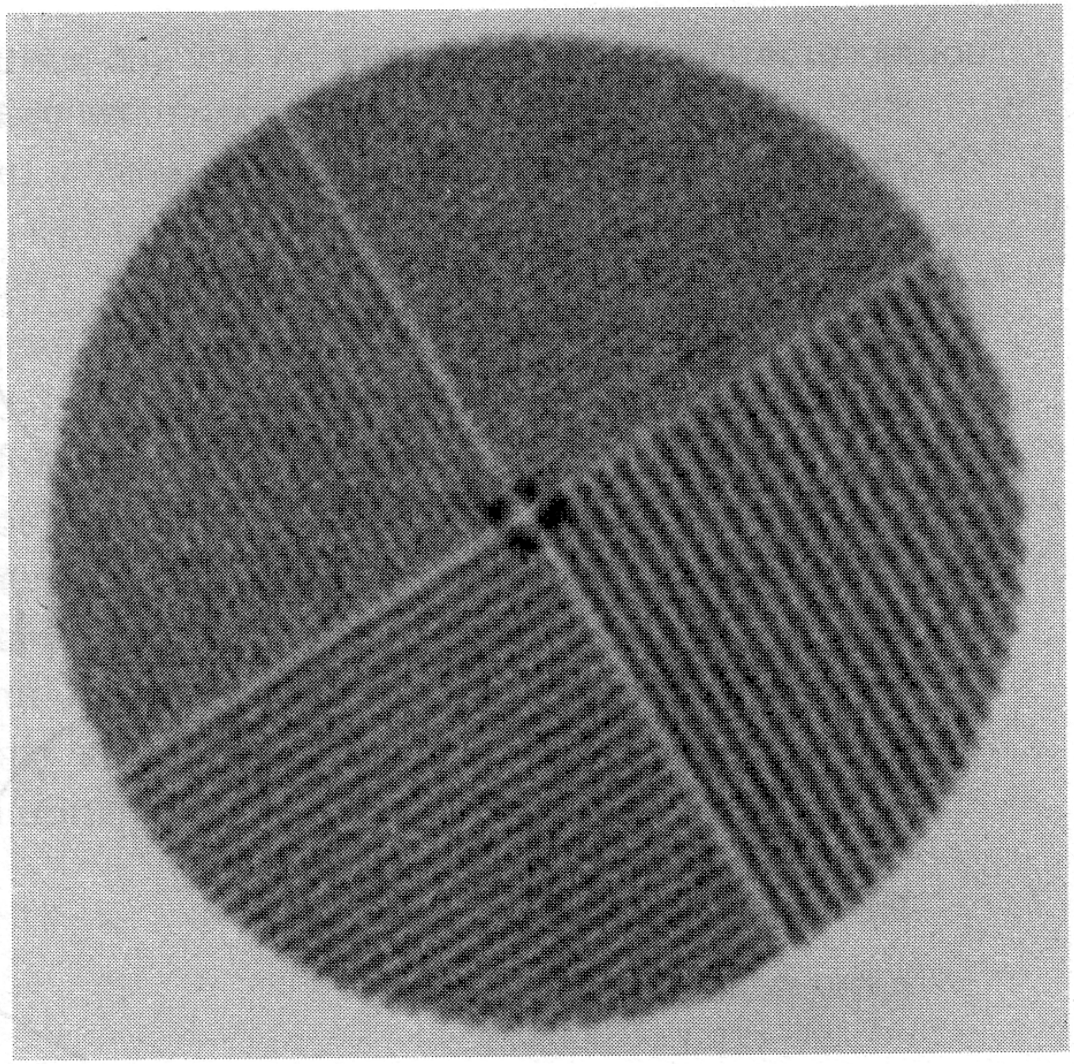
***Quarterly***: *Measure resolution and sensitivity*

- Resolution measured with Jascszak-phantom (see ACR guidelines in RSNA Physics Modules)
- Sensitivity: Use known activity of Tc-99m in a small vial placed on a LEHR collimator
  Calculated as count rate (less the background) divided by phantom activity.
  Should be >200 counts per minute per $\mu$ci.

***Yearly***: *Looking for slowly varying performance changes*

- Detector alignment: using spirit leveler (important for SPECT)
- Multi-window spatial alignment: verify coordinates for different energy windows coincide (±2 mm)
- Dead time Test: Uses two sources imaged separately and together.
  Target count rate for 20% dead time depends on app and geometry (~20 kcps)
- Rotational uniformity: Flood differences acquired at 0, 90,180 and 270 degrees
  Looking for any major artifacts or count rate differences

# Spatial Resolution Test

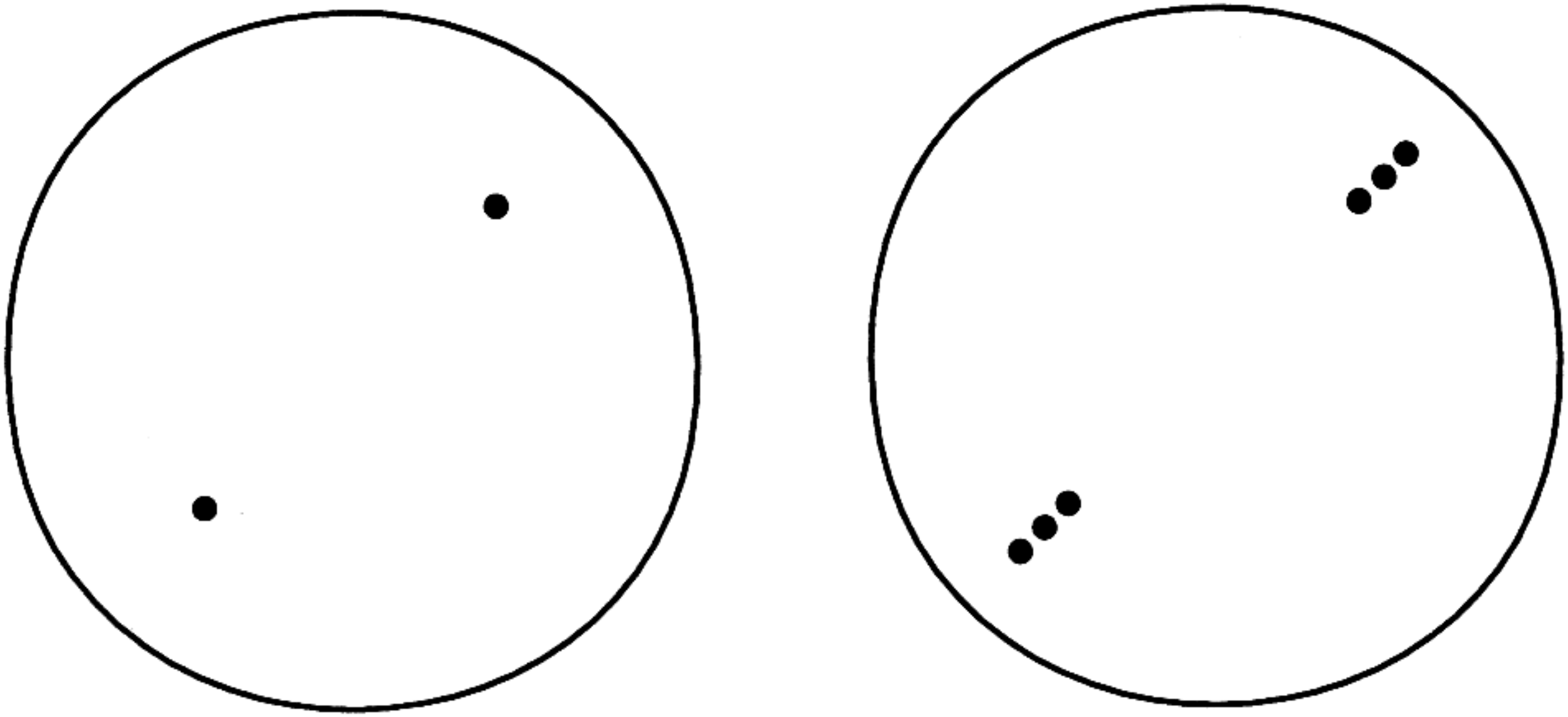


FWHM of LSF = 1.7 x (size of smallest bar resolved)

From: The Essential Physics of Medical Imaging (Bushberg, et al)

# Multienergy spatial registration

(e.g., Ga-67 (93-, 185-, and 300 keV) gamma rays)



properly adjusted | improperly adjusted

From: The Essential Physics of Medical Imaging (Bushberg, et al)

# The Scintillation Camera: Image Acquisition

# Image Acquisition

- Frame mode (data stored as an image)
  - static
    - single image acquisition
    - can have multiple energy windows
  - dynamic
    - series of images acquired sequentially
  - gated
    - repetitive, dynamic imaging
    - used for cardiac imaging
- List-mode (data stored event by event)
  - time stamps are included within data stream
  - allows for flexible post-acquisition binning
  - can result in very large data files

# Gated Acquisition

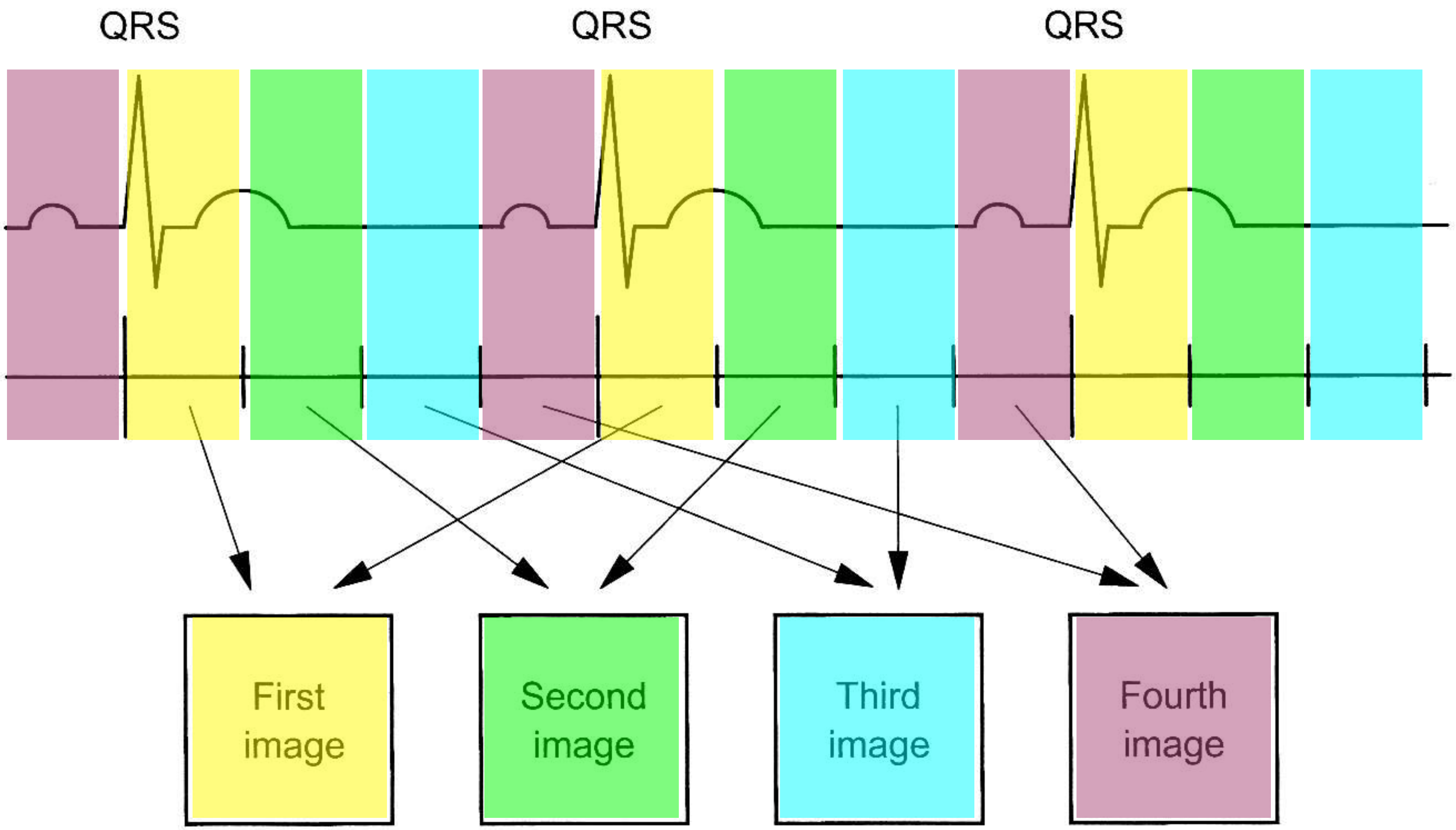



**FIGURE 21-22.** Acquisition of a gated cardiac image sequence. Only four images are shown here. Sixteen to 24 images are typically acquired.

From: The Essential Physics of Medical Imaging (Bushberg, et al)

# Region of Interest (ROI) and Time-Activity Curves (TAC)

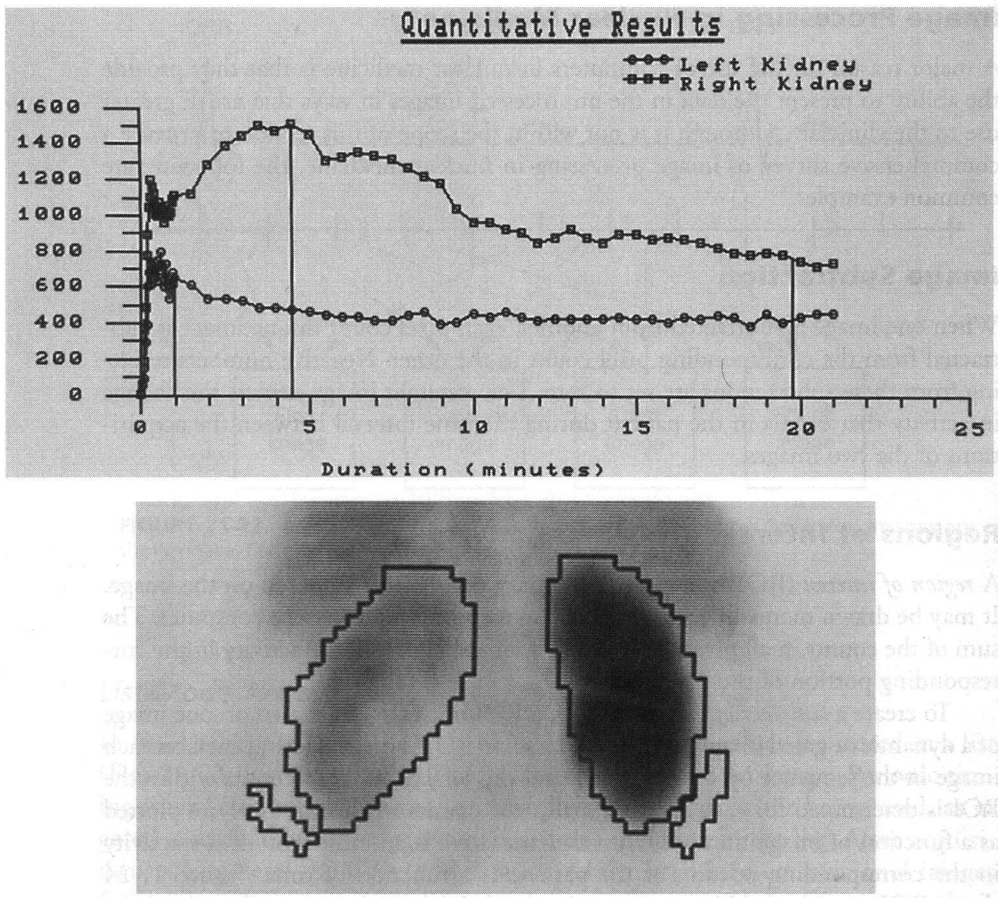



**FIGURE 21-24.** Regions of interest (ROIs) **(bottom)** and time-activity curves (TACs) **(top)**.

From: The Essential Physics of Medical Imaging (Bushberg, et al)

# Raphex Question

**D81.** A cold spot artifact appears in a scintillation camera image. The artifact could be caused by all of the following ***except:***

A. The camera is incorrectly peaked for the radionuclide in the study.
B. The photomultiplier tube is defective.
C. The patient is wearing metallic jewelry.
D. An out-dated uniformity correction is used.
E. The wrong collimator was used.

# Raphex Answer

**D81. E**

The wrong collimator would increase septal penetration and increase or decrease camera sensitivity, but could not produce a cold spot in the image.

# Raphex Question

2-4. In nuclear medicine imaging, match the following quality control procedures with the relevant choice:

a. *Gamma camera resolution*
b. *Gamma camera field uniformity*
c. *Photopeak window of the pulse height analyzer*

2. *Checked daily using a uniform flood source.* _____
3. *Checked daily by placing a small amount of a known source of radioisotope in front of the camera.* _____
4. *Checked weekly using a bar phantom.* _____

# Raphex Answer

1. *Checked daily using a uniform flood source. __b___*
2. *Checked daily by placing a small amount of a known source of radioisotope in front of the camera. _c____*
3. *Checked weekly using a bar phantom. _a____*