



Chapter 7

Gamma-camera and measurement devices

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EPFL - PHYS 455

Introduction to Medical Physics

Learning objectives

- Explain the main **components** of a **gamma-camera/SPECT device** and its **functioning**.
- Explain the **working principle** of different **dosimetric devices** (activimeter, dose rate meter, spectrometer, contamination monitor).
- Identify the different fields of **clinical** and **radiological protection applications**.

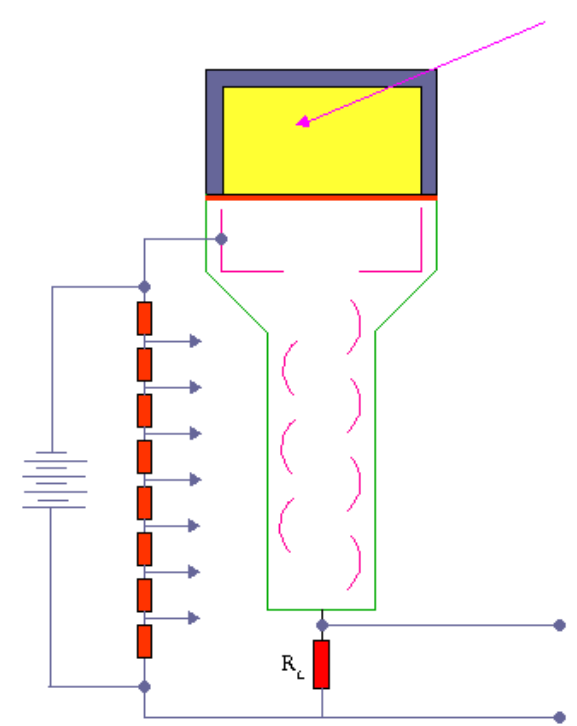
Part 1 – Dosimetric devices

How to choose an instrument?

- It depends on:
- Quantity to measure:
 - Dose, activity, presence of contamination, ...
- Type of decay:
 - alpha, beta, gamma, neutrons, ...
- Type and purpose of the measure:
 - geometry, instantaneous information, individual measurement, estimation vs. precise quantification,
- Other considerations: portability, duration, measurement conditions (e.g. background noise).

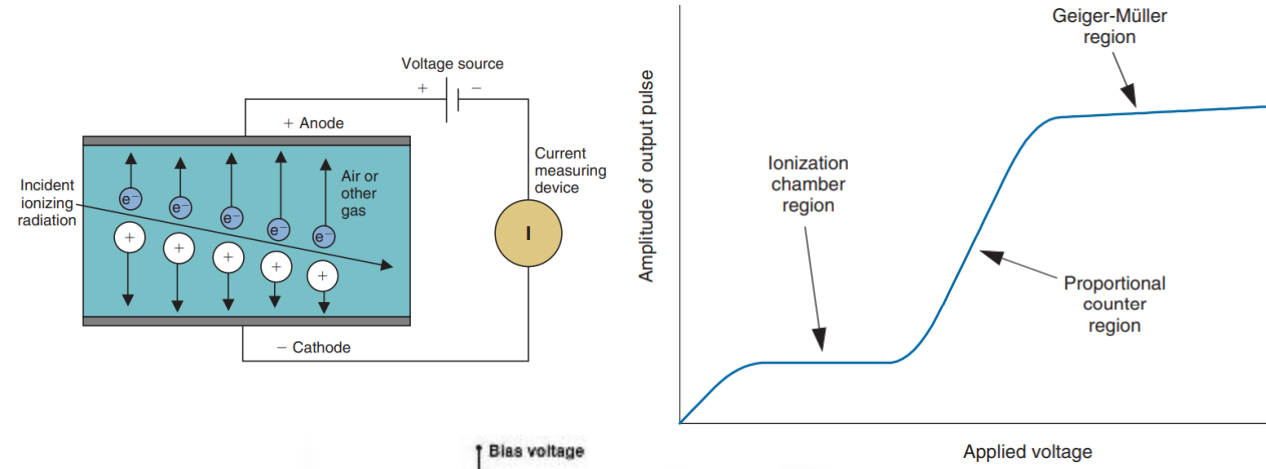
Detectors

- Quantify (and sometimes characterize) ionizing radiation.
- Ionization: process by which electrons are removed from or added to an electrically neutral atom, that becomes a charge carrier.
- An instrument that measures ionizing radiation generally measures the electric current generated by ionization inside the detector.
- The available radiation information is not the same for all types of detectors.
- Different detectors have different fields of application.

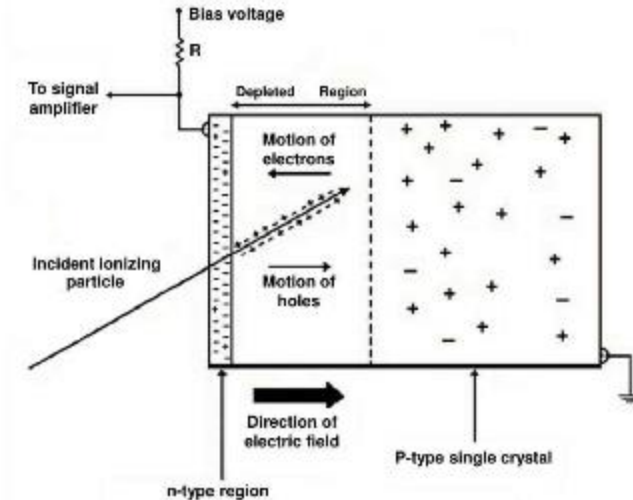


1. Gas detectors

- Ionization chambers
- Proportional counters
- Geiger-Müller counters

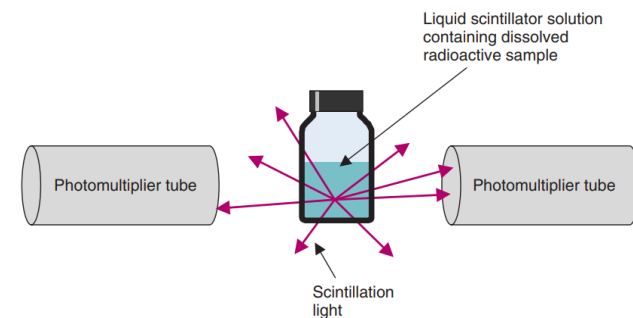
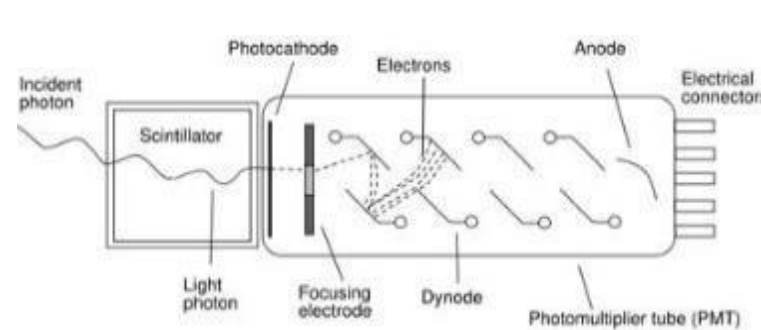


2. Semiconductor detectors



3. Scintillation detectors

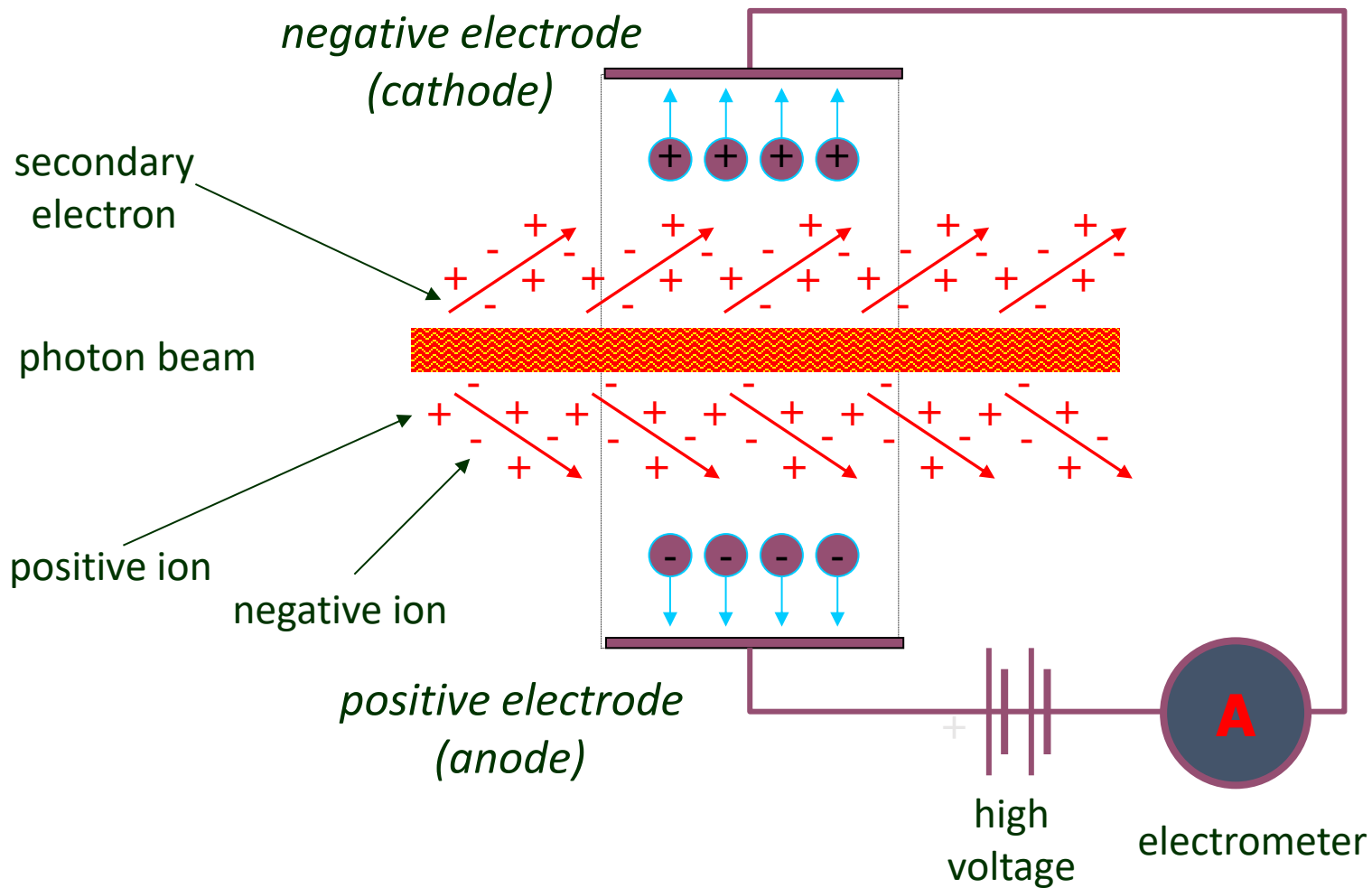
- Inorganic
- Organic



Gas detectors – operating principle

- A volume of gas is contained between two electrodes (anode and cathode) which are kept at different voltages.
- An electric field is thus established between the electrodes.
- The gas filling the detector is an insulator, and there is normally no electrical current flowing between the electrodes.
- When radiation interacts with the gas, it ionizes it (strips off electrons).
- Ionization can be direct or indirect (e.g. generated by secondary electrons).
- Under the influence of the electric field, negatively charged electrons are attracted to the anode (+), while the positively charged atoms of the ionized gas are attracted to the cathode (-).
- This creates an electric current, which can be detected (sometimes after amplification).

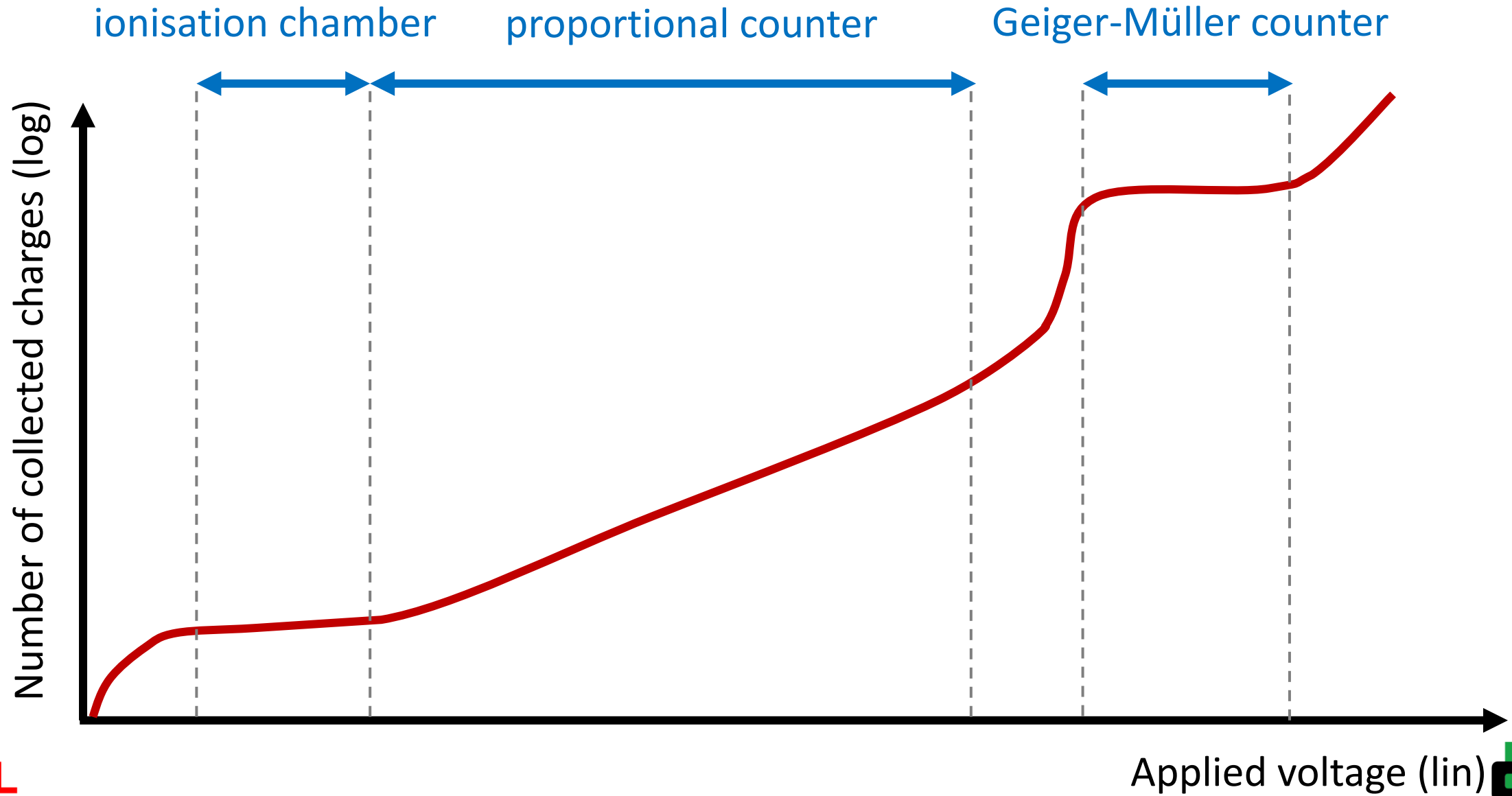
Gas detectors – operating principle



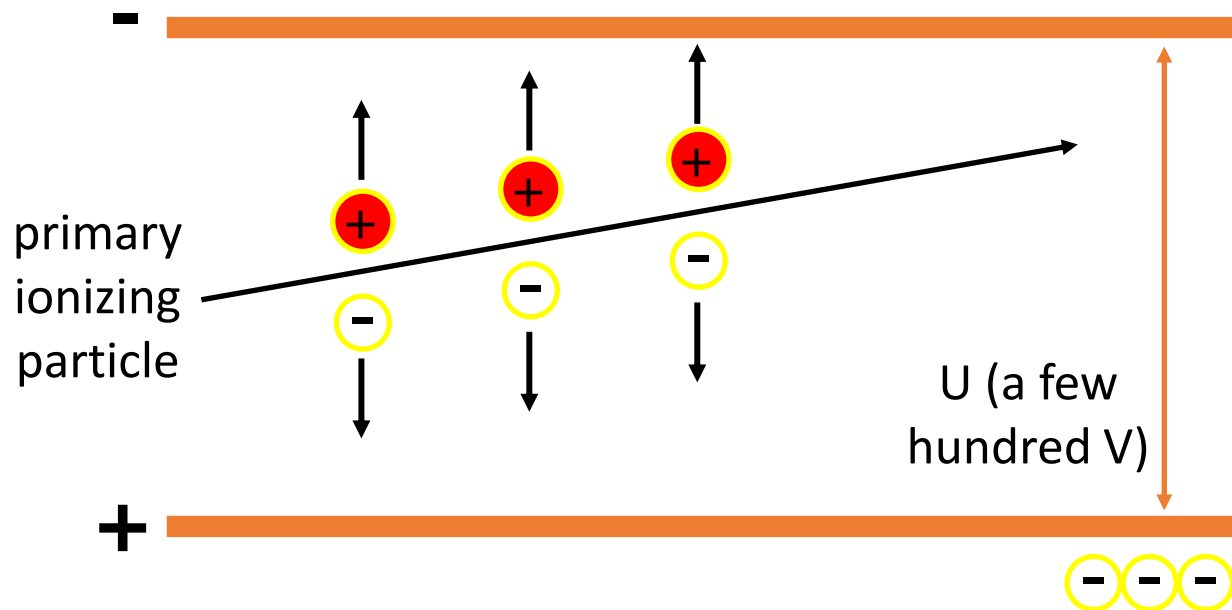
And if the voltage is **too low**?

Recombination between ions (+) and electrons (-), no charge collection, no signal.

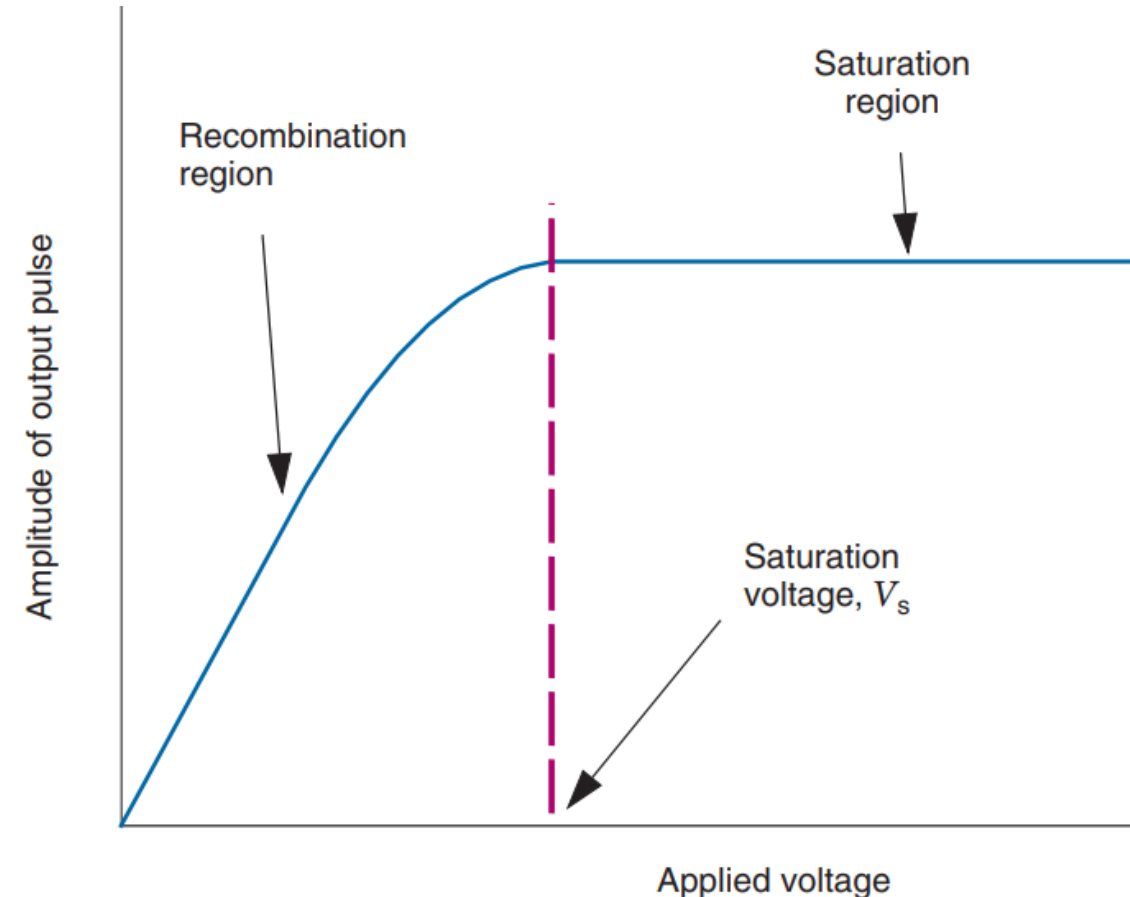
Gas detectors – types



Gas detectors – ionization chamber

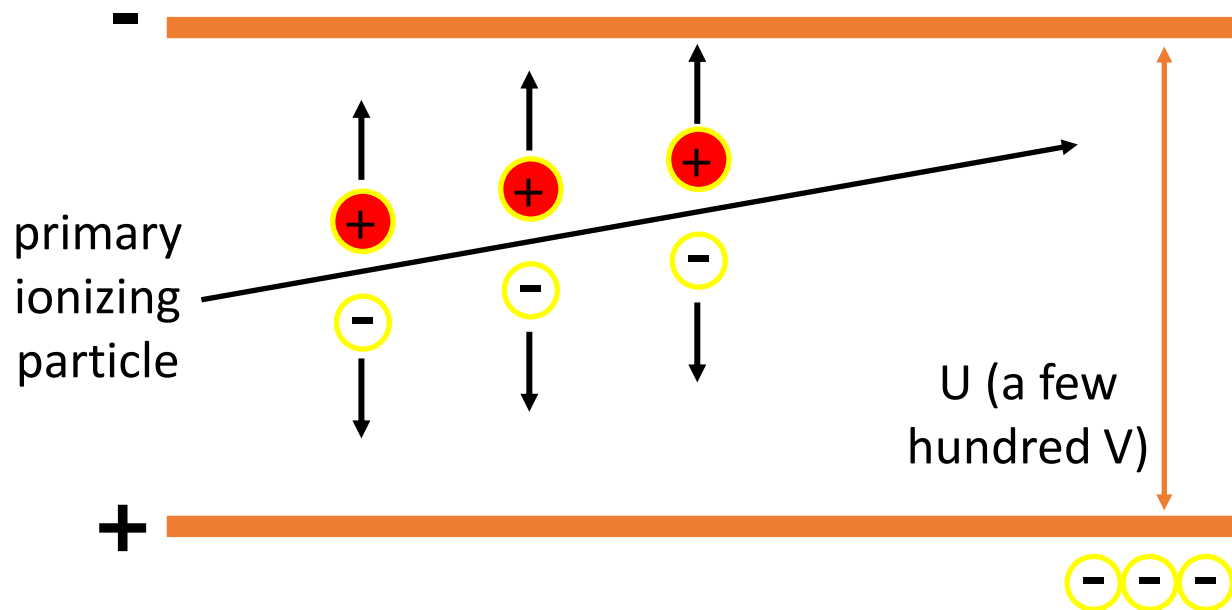


Fast enough charge collection to avoid recombination in the volume



- Absorbed dose measurement
- **Activimeter**

Gas detectors – ionization chamber



Fast enough charge collection to avoid recombination in the volume

Principle

- Measurement of the primary ionization produced in a gas

Characteristics

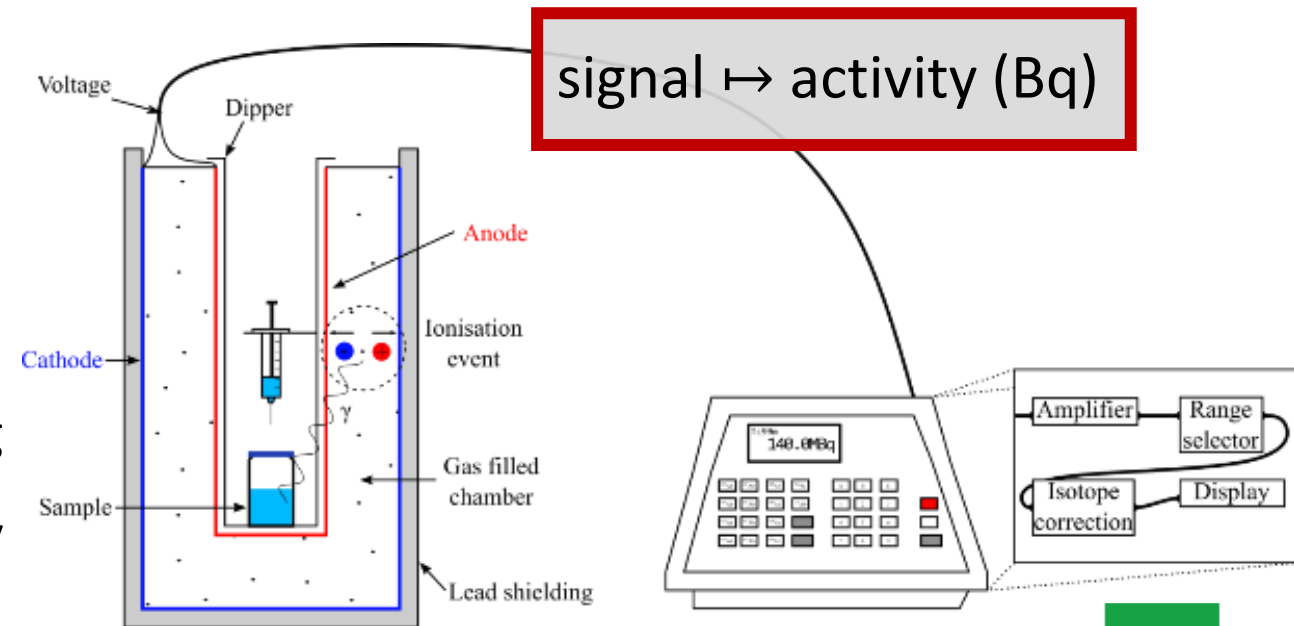
- Recombination at low voltages
- Global measurement (no spectrum)
- Low sensitivity
- Largely independent of radiation type and energy

Typical applications

- Absorbed dose measurement
- **Activimeter**

Gas detectors – ionization chamber (application)

- Clinical dosimetry, field strength meters, **activimeters**.
 - Gives information about the activity (Bq) of a sample (e.g. syringe, vial).
 - To do this, a conversion factor between current and activity must be applied.
 - The factor depends on the radionuclide and is strongly influenced by the measurement geometry (vial factor \neq syringe).
 - Sealed chamber containing pressurized Ar (to limit the influence of ambient pressure).
 - ⚠ does not identify a radionuclide (no info on energy!)
 - Always specify the radionuclide being measured and respect the geometry used during the calibration!

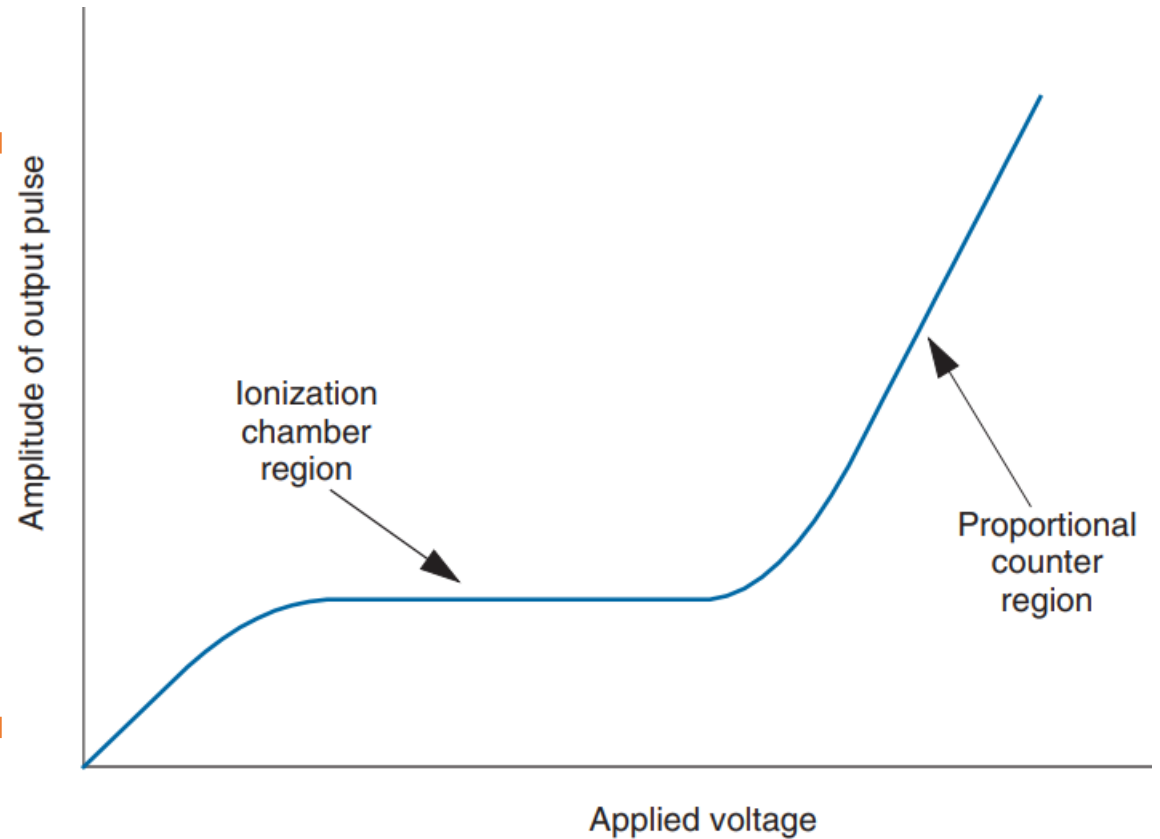
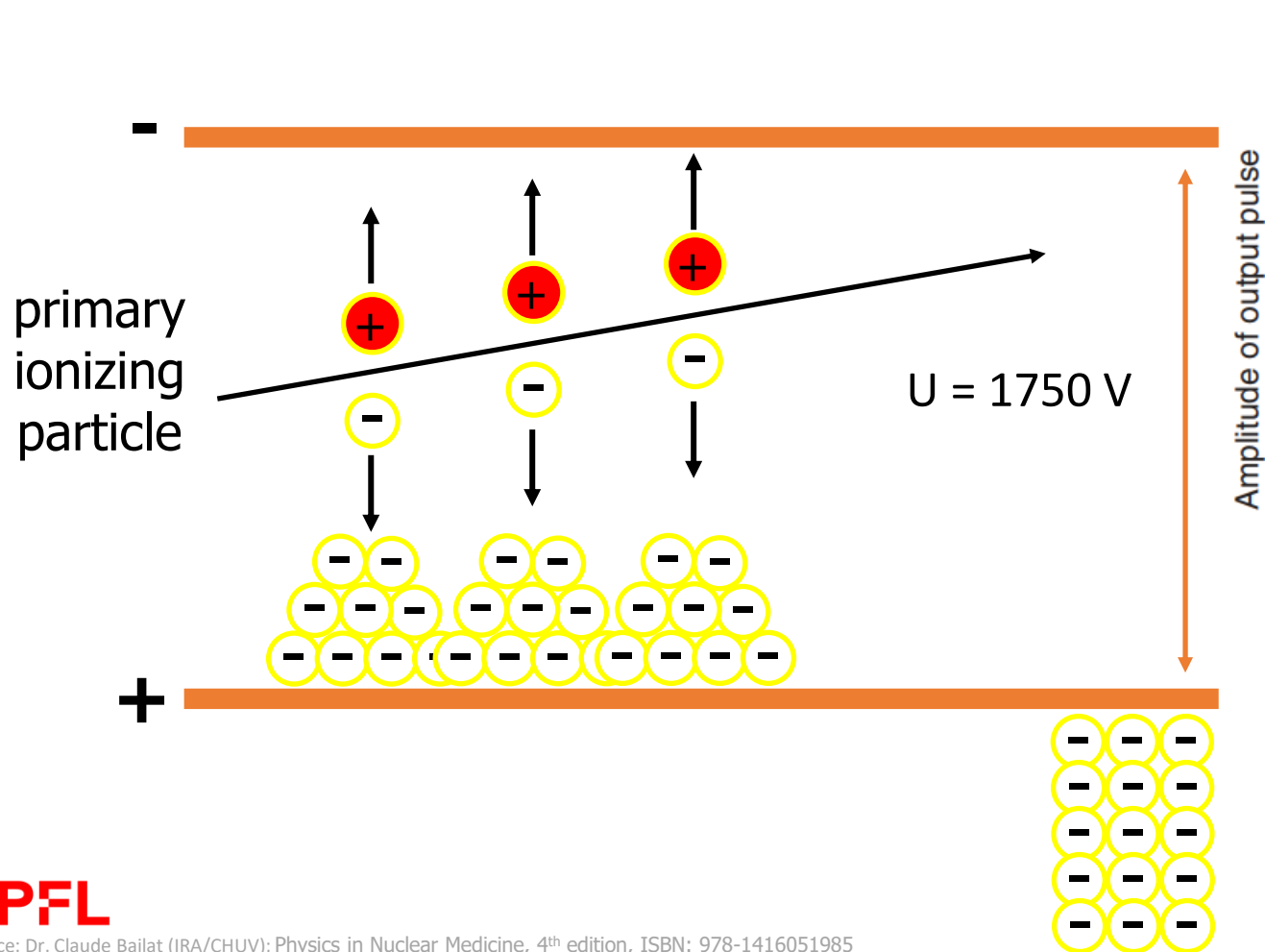


Gas detectors – proportional counters

- Ionization chamber: the voltage between anode and cathode is just sufficient to collect the charges released by the direct action of radiation.
- If the voltage is increased, the electrons generated by the ionization events themselves gain enough energy to create other ionizations within the gas, and so on.
- Amplification process (*Townsend avalanche*) in gas → 1 ionization = 1 avalanche.
- Quantifiable by the amplification factor, which increases with the voltage applied between the electrodes (up to a factor of 10^6).
- Total charge ionization energy (a few eV) x amplification factor.
- The charge produced (and hence the current and signal detected) is proportional to the total energy deposited in the detector by the ionizing radiation.

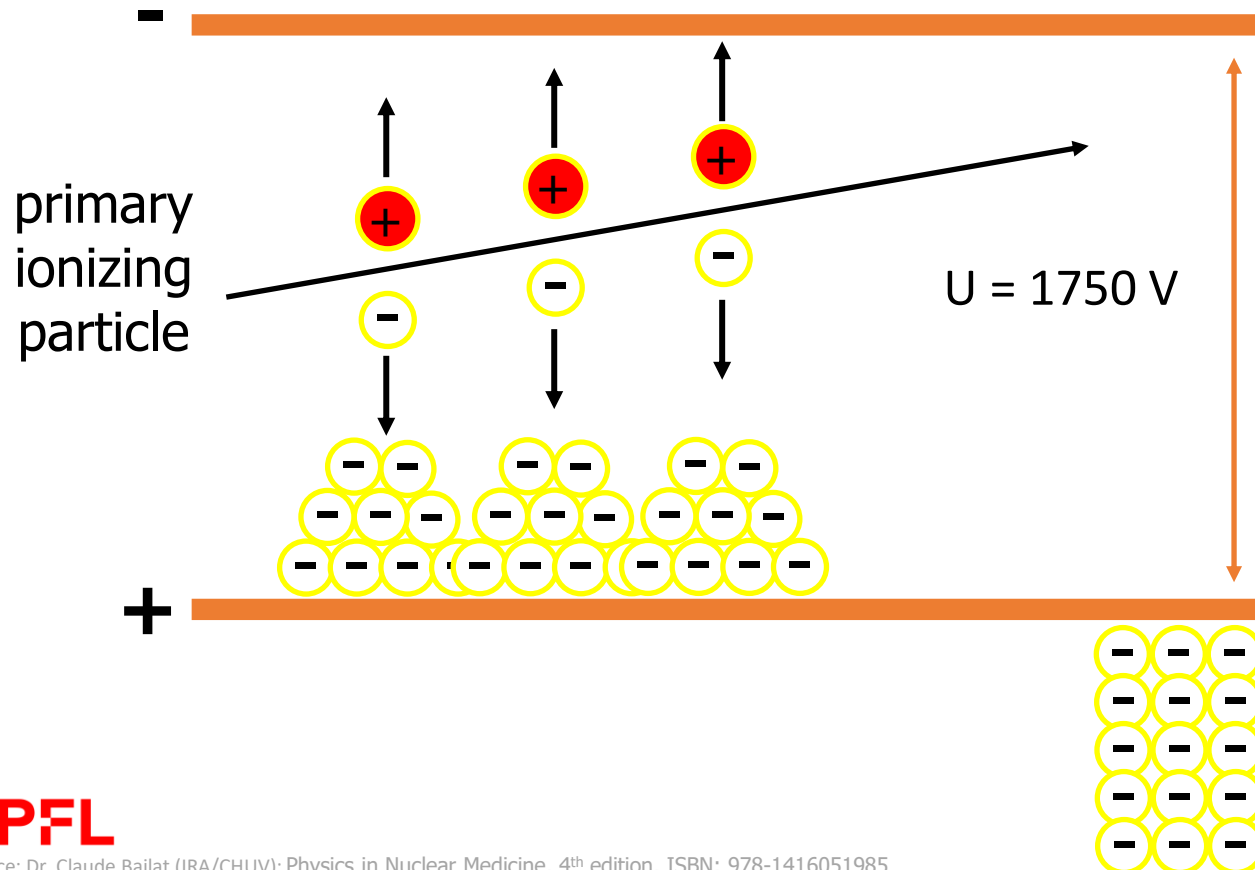
Gas detectors – proportional counters

- True proportionality



Gas detectors – proportional counters

- True proportionality



Principle

- Measurement of secondary ionization produced by acceleration of primary charges produced

Features

- Relatively high voltage
- Output signal \propto to primary charge produced
- Pulse measurement
- High sensitivity

Typical applications

- Surface contamination monitor

Gas detectors – proportional counters

- Applications: field measurement, **surface contamination monitor**, Low-energy X-ray detection and spectrometry
- Count rate proportional to surface activity (Bq/cm^2)
- Units: “counts per second” or “impulses per second” \rightarrow “cps”, “ips” or “s⁻¹”.
- A certified laboratory (e.g. IRA, PSI) delivers a certificate:
 - The instrument is exposed to a set of sources with a known surface activity (Bq/cm^2)
 - The count rate corresponding to the known surface activity is recorded in the verification certificate for each radionuclide used.
 - The verification certificate gives the correspondence between the cps (displayed by the instrument) and the surface contamination (Bq/cm^2).



Quiz - Gas detectors – proportional counters

- During the handling of I-124 (outside a controlled area), a drop of solution falls on the ground.
- The contamination monitor indicates 12 counts per second (net).
- Is this contamination above the limit set by the Swiss Radiation Protection Ordinance (i.e. 10 Bq/cm² for I-124)?

Quiz - Gas detectors – proportional counters

Radioprotection. O



814.501

Nucléide	Période	Mode de désintégration / rayonnement	Grandeurs d'appréciation					Limite de libération	Limite d'autorisation	Valeurs directrices		Nucléide de filiation instable
			e_{inh} Sv/Bq	e_{ing} Sv/Bq	h_{10} (mSv/h)/ GBq à 1 m de distance	$h_{0,07}$ (mSv/h)/ GBq à 10 cm de distance	$h_{c,0,07}$ (mSv/h)/ (kBq/cm ²)	LL Bq/g	LA Bq	CA Bq/m ³	CS Bq/cm ²	
1	2	3	4	5	6	7	8	9	10	11	12	13
I-123	13.27 h	ec/ph	1.10E-10	2.10E-10	0.043	400	0.3	1.E+02	5.00E+07	8.00E+04	30	→ Te-123
I-124	4.1760 d	ec, β ⁺ /ph	6.30E-09	1.30E-08	0.170	300	0.5	1.E+01	8.00E+05	1.00E+03	10	
I-125	59.400 d	ec/ph	7.30E-09	1.50E-08	0.033	4	<0.1	1.E+02	7.00E+05	1.00E+03	10	
I-131	8.02070 d	β ⁻ /ph	1.10E-08	2.20E-08	0.062	1000	1.4	1.E+01	5.00E+05	8.00E+02	3	→ Xe-131m

Les valeurs figurant dans la colonne "Val. arrondies" sont des valeurs pratiques en accord avec les tolérances de l'ordonnance sur les instruments de mesure des rayonnements ionisants (OIMRI).

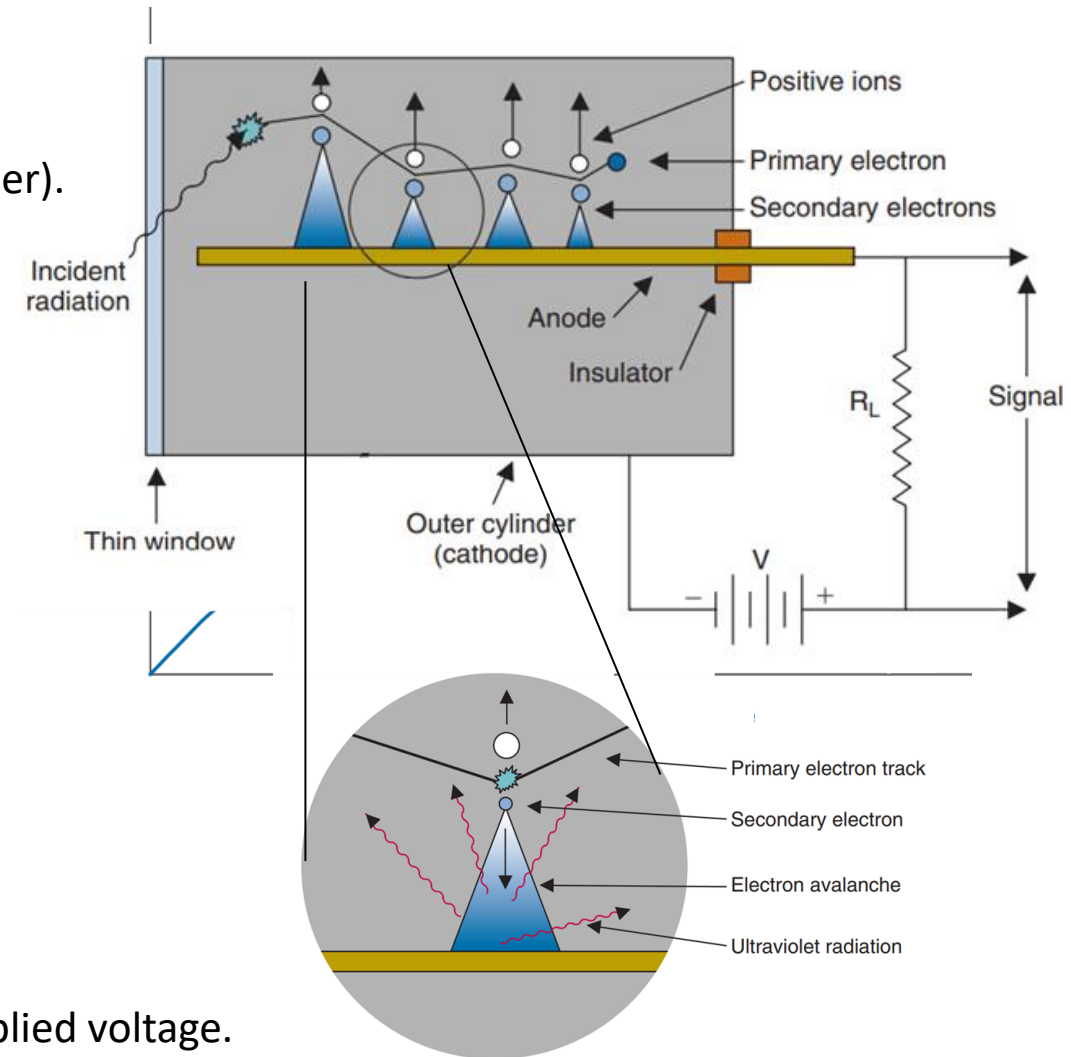
Nuclide	Valeur directrice CS [Bq cm ⁻²]	Valeur limite [cps]	
		Vérification	Val. arrondies
C-11	3	56	60
N-13	3	64	60
O-15	3	64	60
F-18	3	54	50
Sc-44	3	55	60
Co-55	3	45	50
Cu-64	10	100	100
Ga-68	3	57	60
Y-86	10	99	100
Zr-89	10	7.5	8
Tc-99m	30	43	40
I-124	10	23	20

certificat:
I-124 = 20 cps

I-124 = 10 Bq/cm²

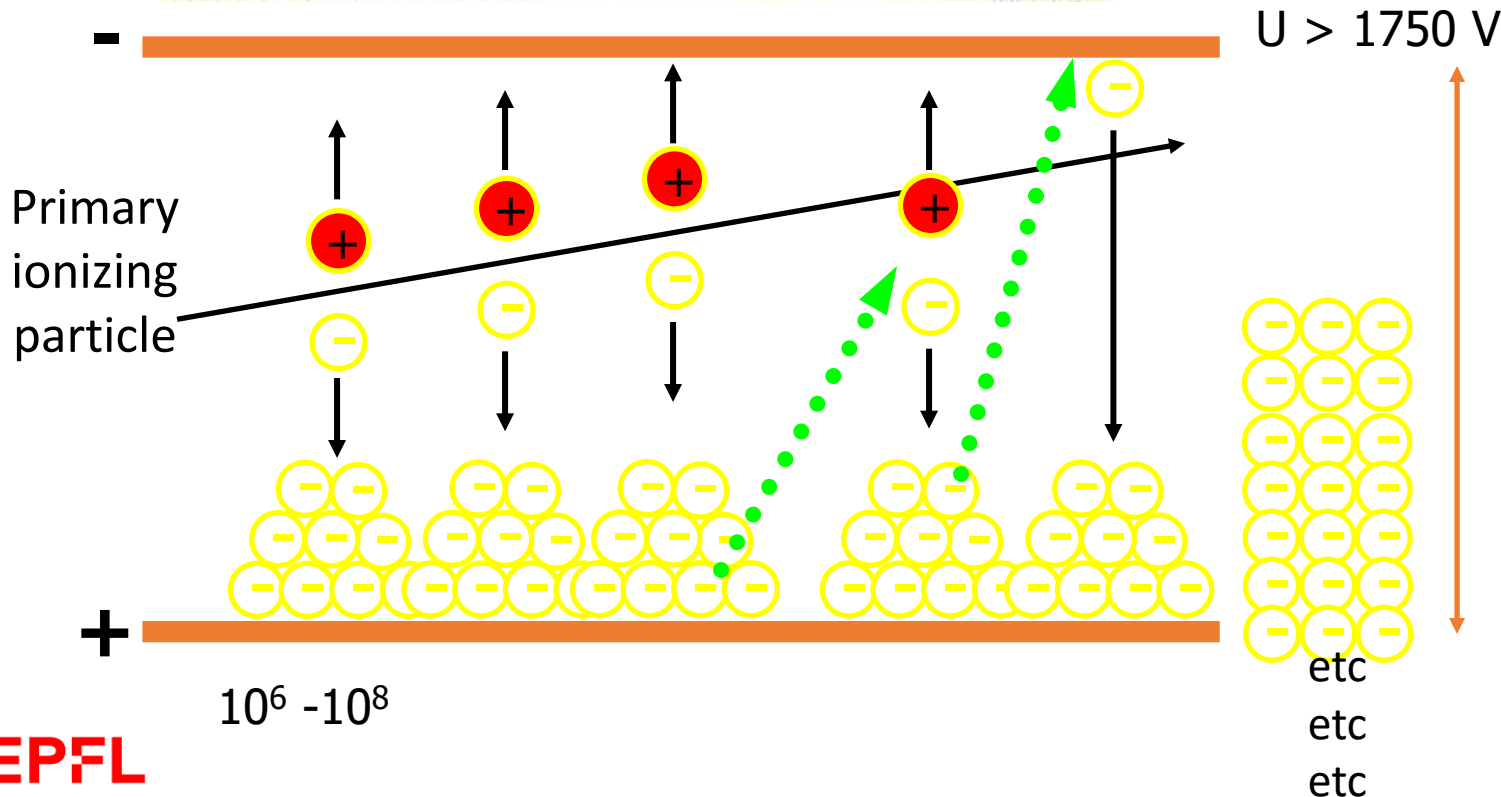
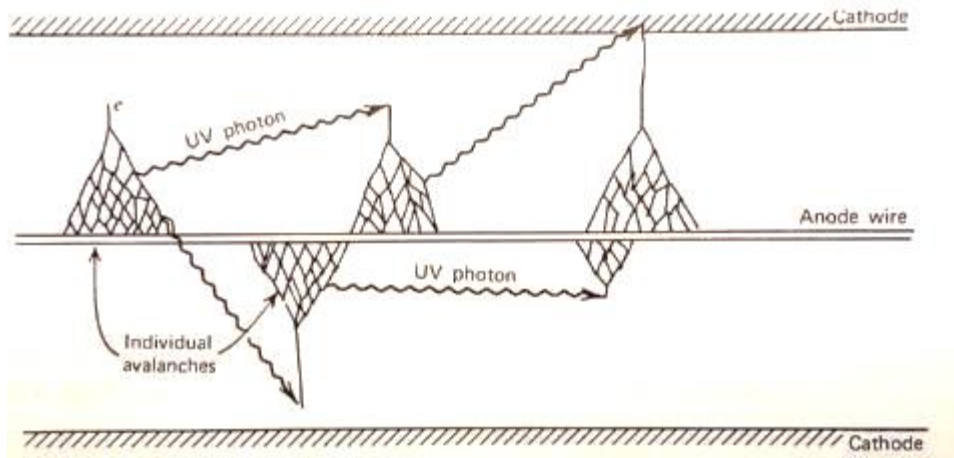
Gas detectors – Geiger-Müller counters

- Purpose: to maximize signal amplification (up to 10^{10}).
- Big voltage \neq between anode (+, central rod) and cathode (-, outer cylinder).
- Sealed cylinder, rare gas mixture + *quenching gas*.
- During ionization, e⁻ is attracted to the anode + amplification process (cf. proportional counters).
- The e⁻ can also induce the excitation of gas molecules by collision.
- Excited molecules return to their ground state, releasing energy in the form of visible or UV light.
- UV photons \rightarrow photoelectric effect with gas or anode material.
- The electron thus released can generate a new avalanche of events.
- The amplitude of the collected signal is large and independent of the applied voltage.
- Individual events can be counted.
- The amplitude of the outgoing signal is constant \rightarrow no energy discrimination!



Incoming radiation creates ion pairs via direct ionization and secondary δ rays. These pairs trigger an avalanche multiplication process, further amplified by UV photon emission. The process self-terminates when positive ions accumulate near the anode, reducing the local electric field due to space charge effects.

Gas detectors – Geiger-Müller counters



Principle

- Maximum signal amplification (avalanche)

Features

- Signal independent of primary charge produced
- Pulse measurement
- Large dead time

Applications

- Ambient dosimeter
- Robust device

Gas detectors – Geiger-Müller counters

Features of GM counter:

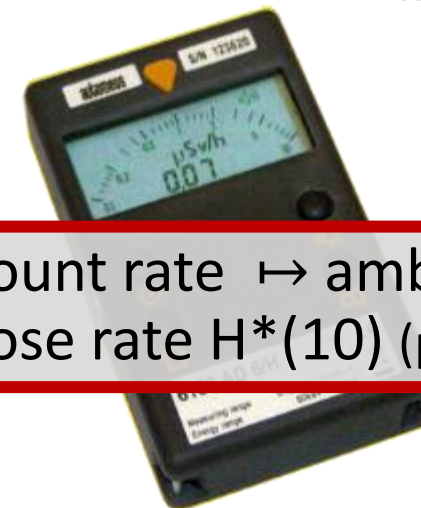
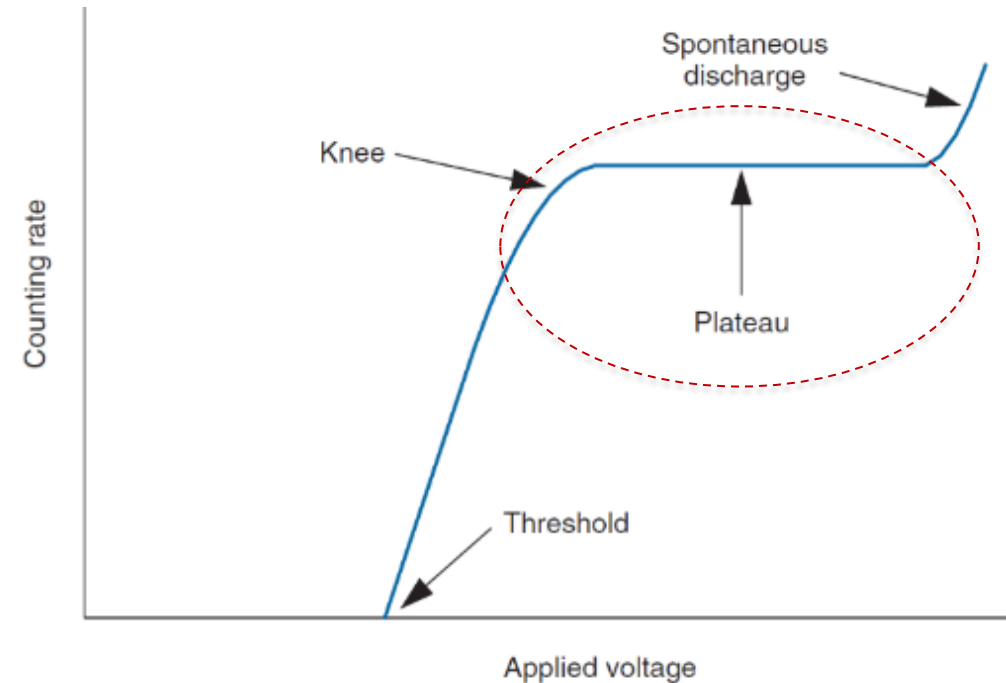
- High voltage
- Output signal independent of the primary ionization
- Spectroscopy not possible
- Dead-time → *Quenching gas* stops the avalanche.

Application domain: plateau

... And if you increase the tension even further?

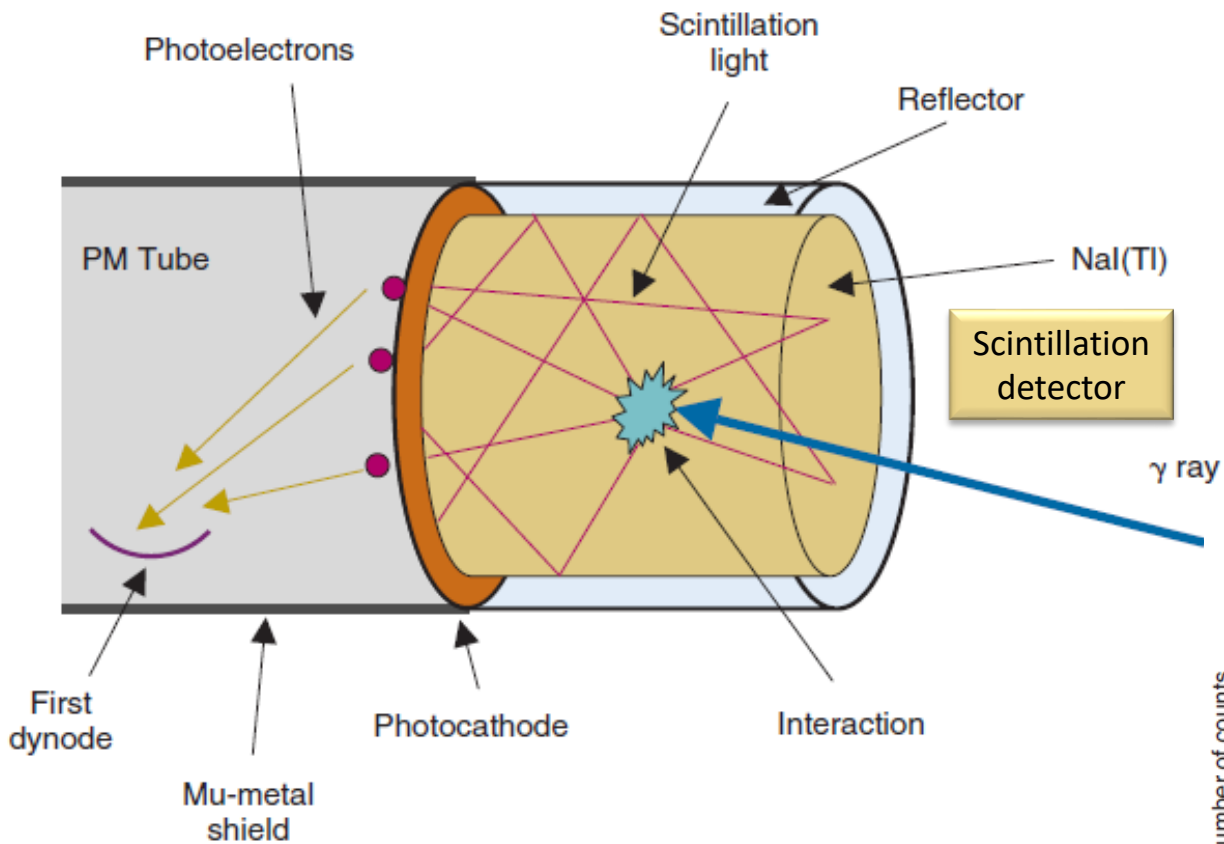
- domain of spontaneous discharge.
- The tension is so high → spontaneous ionizations.
- No useful information in this domain!

Main applications → measurement of the dose rate

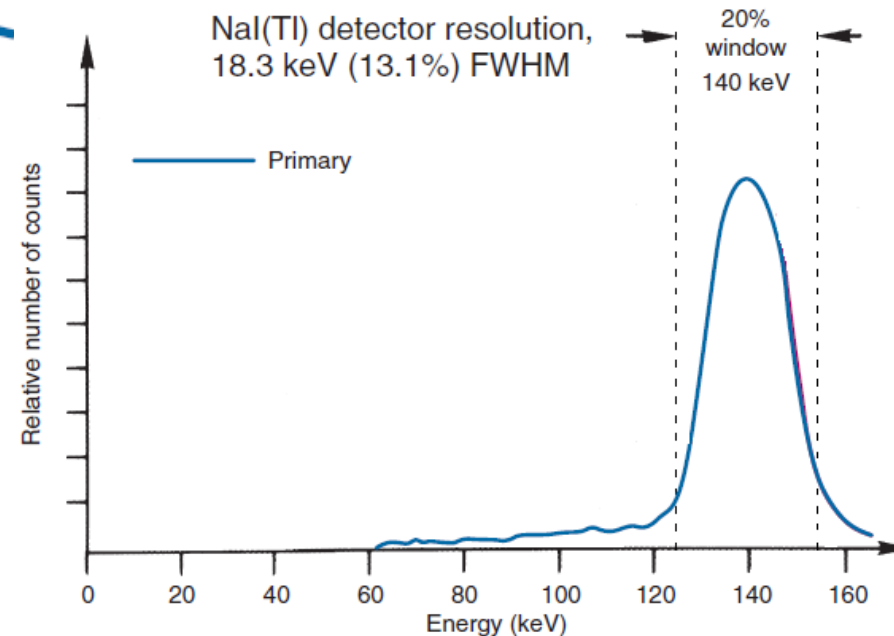


Count rate → ambient equivalent dose rate $H^*(10)$ ($\mu\text{Sv/h}$)

Scintillation detectors



- Scintillation: property of certain materials to emit light when they return to their ground state after interacting with ionizing radiation.
- The amount of light emitted is proportional to the energy of the radiation.
- The light is converted into an electrical signal and then amplified (e.g. photomultiplier tube PMT).



Types of scintillators:

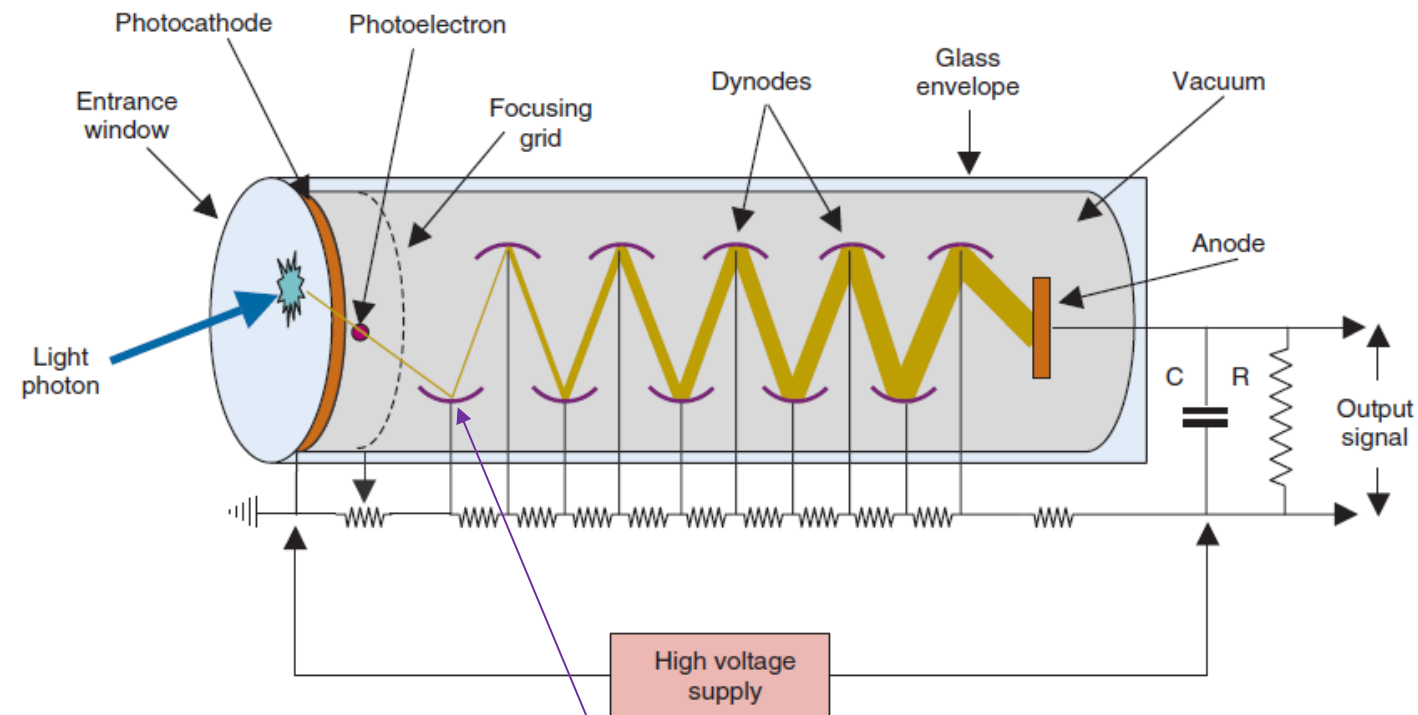
Organic (liquid, plastic)

Inorganic (crystals)

- NaI(Tl), BGO, LSO, LYSO, CsI, LaBr₃(Ce)

Amount of scintillation light \propto deposited energy \rightarrow spectroscopy possible!

Scintillation detectors



electron multiplication factor per dynode: $\times 6$
for a PMT tube with 10 dynodes = 6^{10}

Principle

- Interaction btw radiation and matter
- De-excitation of atoms
- Light emission

Characteristics

- Crystals of NaI (Ti), ZnS, etc...
- Strong variation with radiation energy

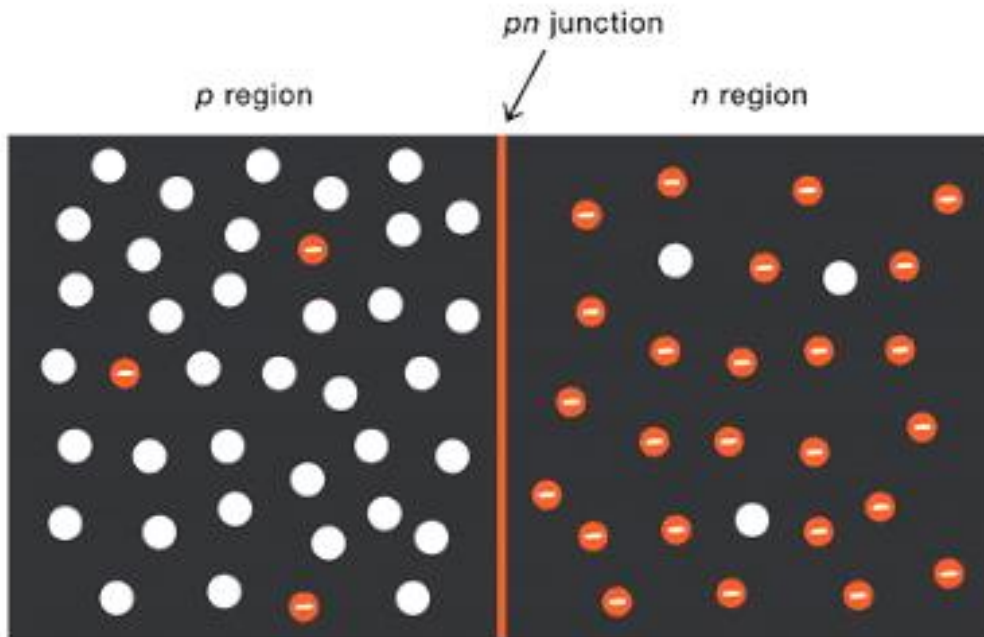
Applications

- Contamination measurement
- Spectro γ
- Liquid scintillation for measuring β

cf. rest of the course for one of its main applications → the gamma camera

Semiconductor detectors

- The materials used (Si, Ge) are not particularly conductive.
 - perfect crystalline structure (covalent) that leaves no room for e- movements.
- Their behaviour can be changed by adding impurities (*doping*).
- N-type (e.g. P, As): 5 e-, they insert themselves into the structure, leaving one e- free (-).
- P-type (e.g. B, Ga): 3 e-, they fit into the structure, leaving a free hole (+).
- Putting together materials of type P and N we obtain:



At the junction, some of the free e- from the n-type move towards the p-type (random thermal migration) → in the p-type, they combine with holes and cancel each other out.

At the junction, some p-type holes (+) move towards the n-type → they combine with free electrons and cancel each other out.

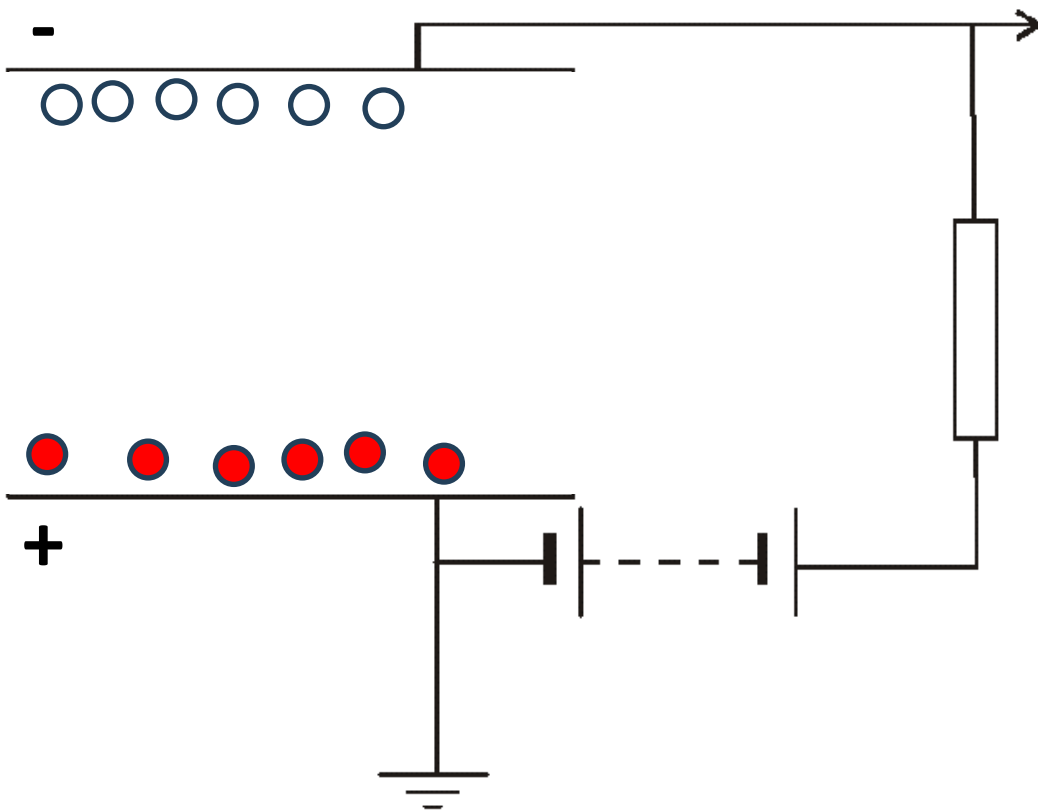
Dopant atoms in the n-type (electron donors) are part of the crystal and cannot move → in the n-type, a region close to the junction becomes "+".

Doping atoms in the p-type (electron acceptors) are part of the crystal and cannot move → in the p-type, a region close to the junction becomes "-".

Depletion layer: region close to the junction which acts to repel mobile charges away from the junction thanks to the electric field these charged regions create.

Semiconductor detectors

● electrons
○ holes



Principle

- Creation of charges in the pn junction of a semiconductor material (solid)
- Current is detected in the form of pulses.
- # of pulses \rightarrow radiation intensity.
- amplitude of the pulse \rightarrow energy deposited

Features

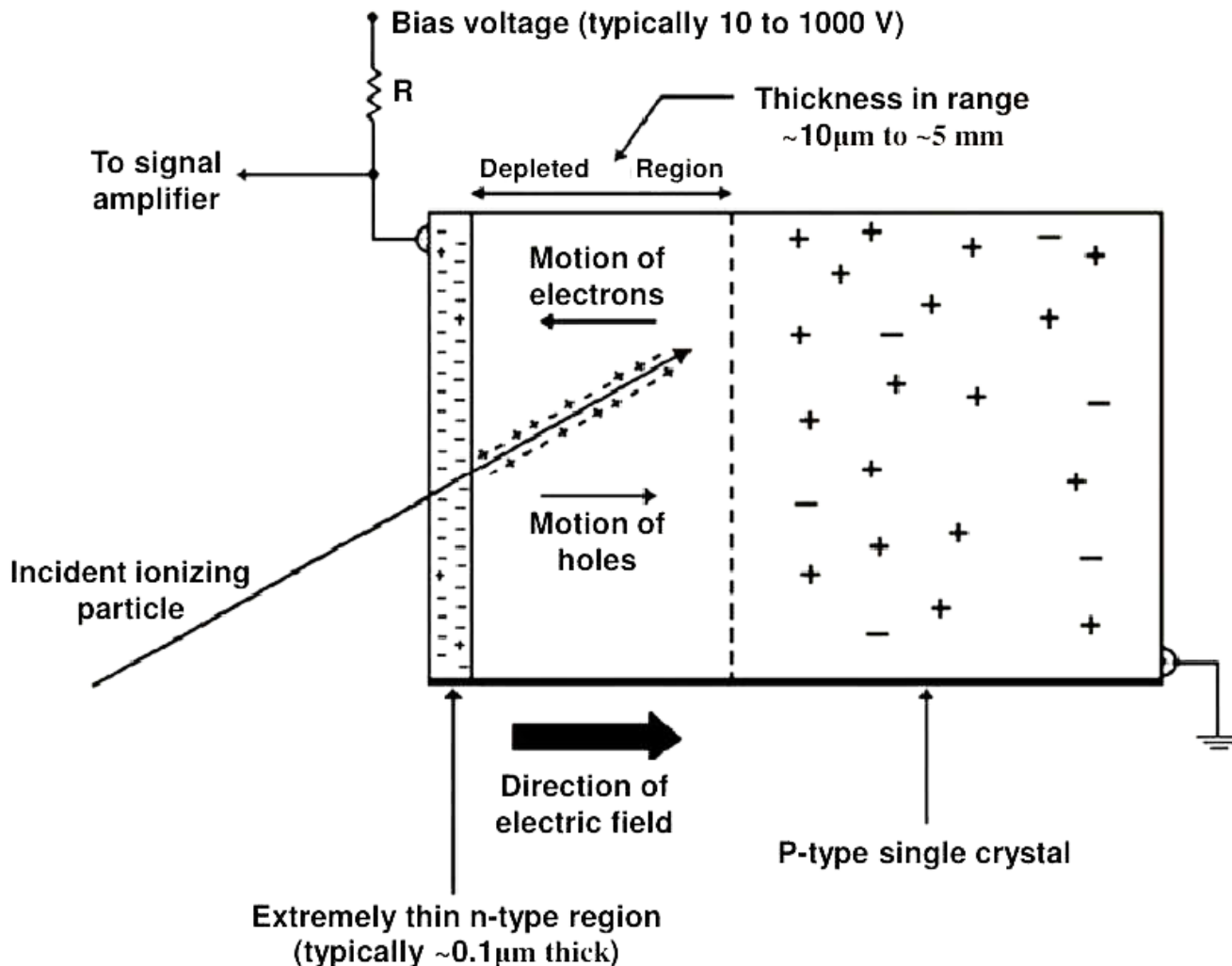
- Small counting volume
- Strong variation of response with radiation energy

Applications

- Diagnostic dosimetry
- High-resolution γ spectro

Semiconductor detectors

- electrons
- holes



Principle

- Creation of charges in the *pn* junction of a semiconductor material (solid)
- Current is detected in the form of pulses.
- # of pulses → radiation intensity.
- amplitude of the pulse → energy deposited

Features

- Small counting volume
- Strong variation of response with radiation energy

Applications

- Diagnostic dosimetry
- High-resolution γ spectro

Semiconductor detectors - applications

- Some semiconductor detectors are excellent options when an unknown γ source needs to be identified (e.g. HPGe, high purity germanium).

→ gamma spectroscopy

- But they also have certain disadvantages (for HPGe):
 - High prices
 - No large crystals (limited efficiency)
 - Fragility
 - Cooling (few portable options) → not the case with CZTs!

Nal (scintillator)



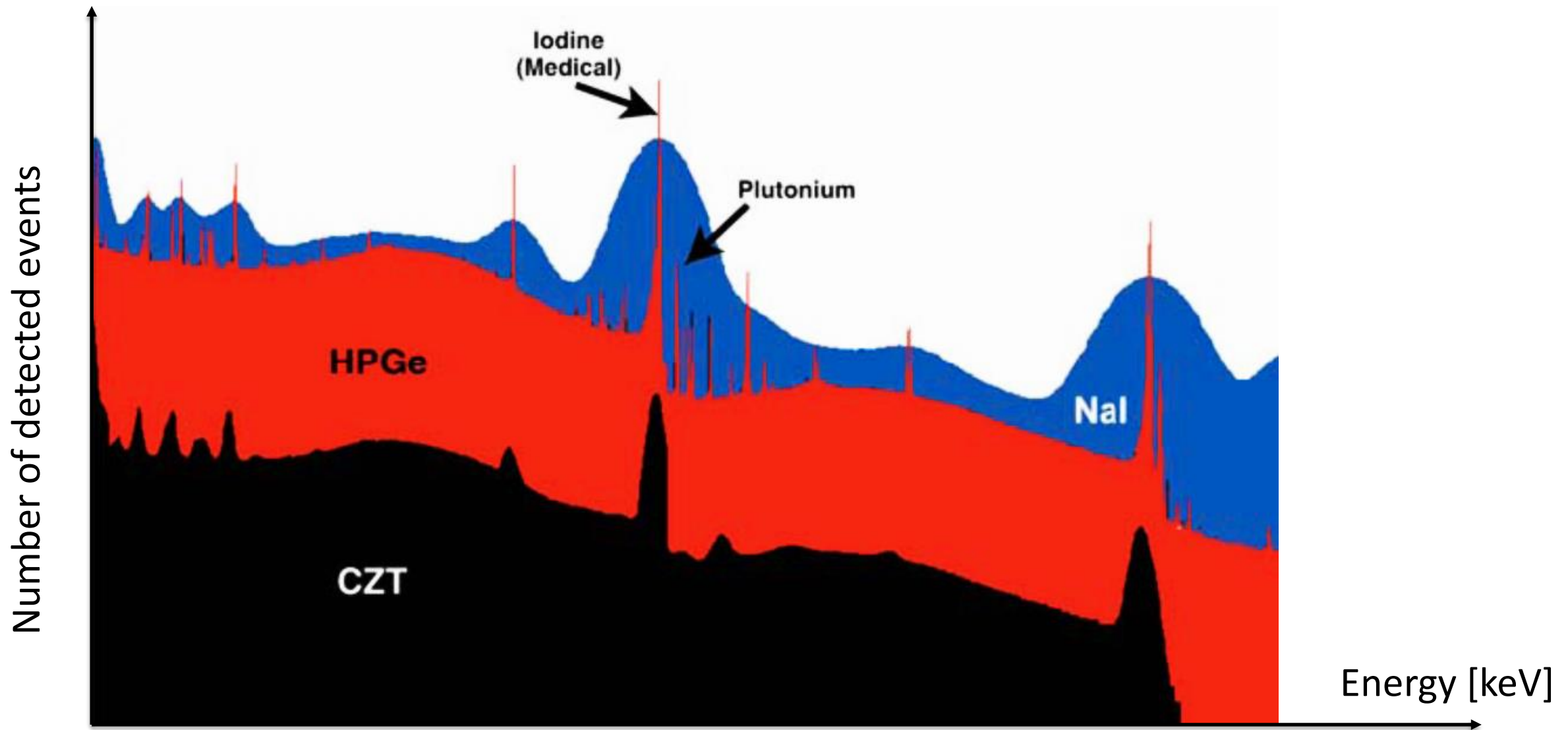
HPGe (semiconductor)



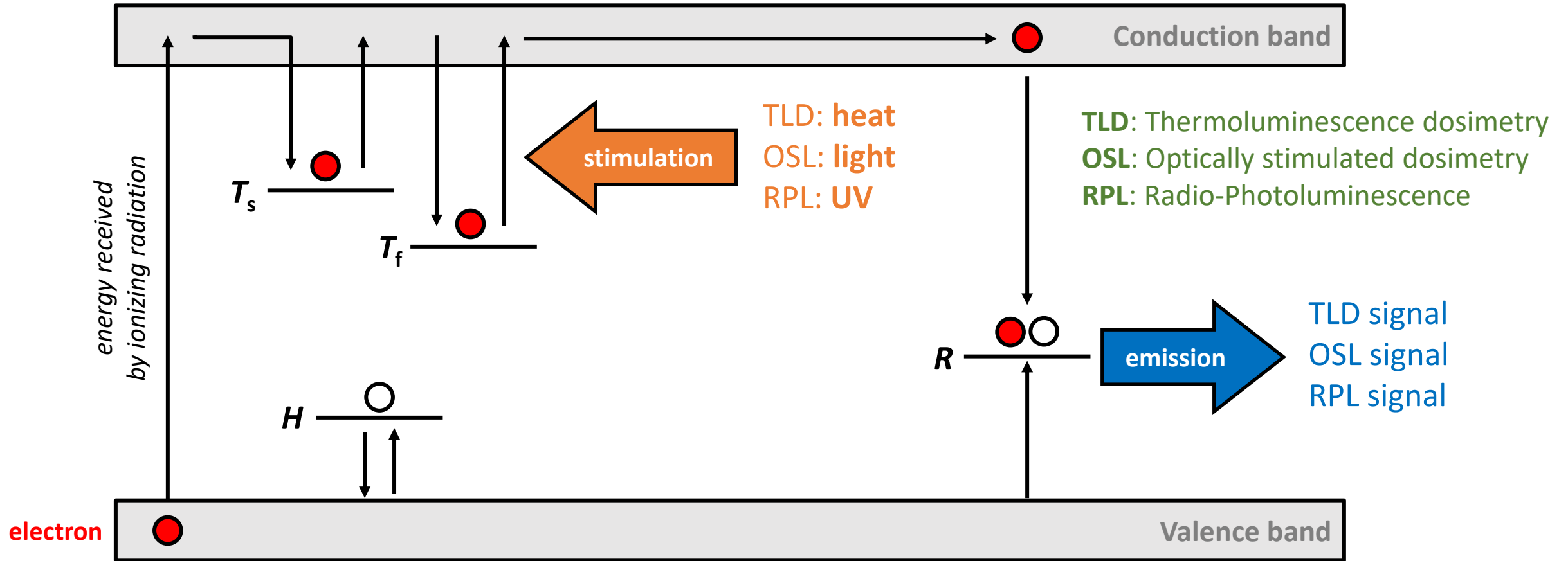
Source: <https://www.isus.de/low-energy-spectrometer>

Source: <https://repository.cern/records/71bjf-gb626>

Semiconductor detectors - applications



Luminescence

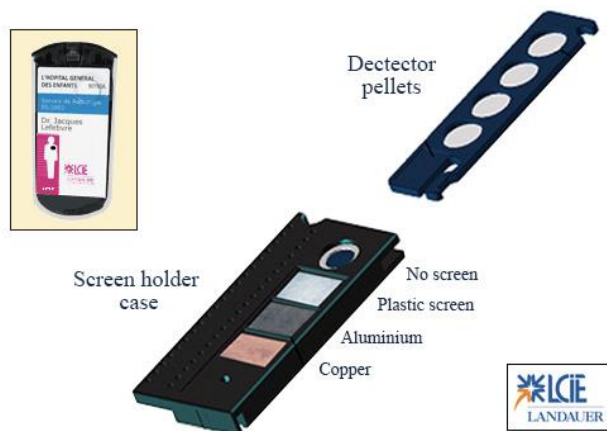


T_s : electron unstable trap (probability of escaping is large)
 T_f : electron storage trap (probability of escaping is negligible)

H : hole trap
 R : recombination center

Luminescence – applications: passive dosimeters

- The material used has imperfections (impurities, mechanical stress, etc.)
- The material is initially in equilibrium (ground state).
- Ionizing radiation excites the e- of the material, which go to a higher (excited) energetic state.
- Most of the e- return to their initial state, but some remain trapped in the material's impurities.
- When energy is supplied (e.g., light or heat), the e- can then be released from the traps.
- The energy released usually manifests itself in the form of (measurable) light.
- The amount of light detected is \propto to the # of trapped electrons ... and thus to the dose.



Passive vs. active dosimeters

Passive:

- Retrospective dose indication.
- Used to measure individual doses.
- Satisfy the requirements set by the Swiss Ordinance on Dosimetry (linearity, energy response, dose range...).
- Have to be read by an accredited laboratory.
- Background is subtracted by the laboratory.
- Typical dose accumulation interval = 1 month.
- Individual doses are recorded in the dosimetric registry.
- Examples :
 - TLD: destructive reading (can only be read once)
 - OSL / RPD : non-destructive reading



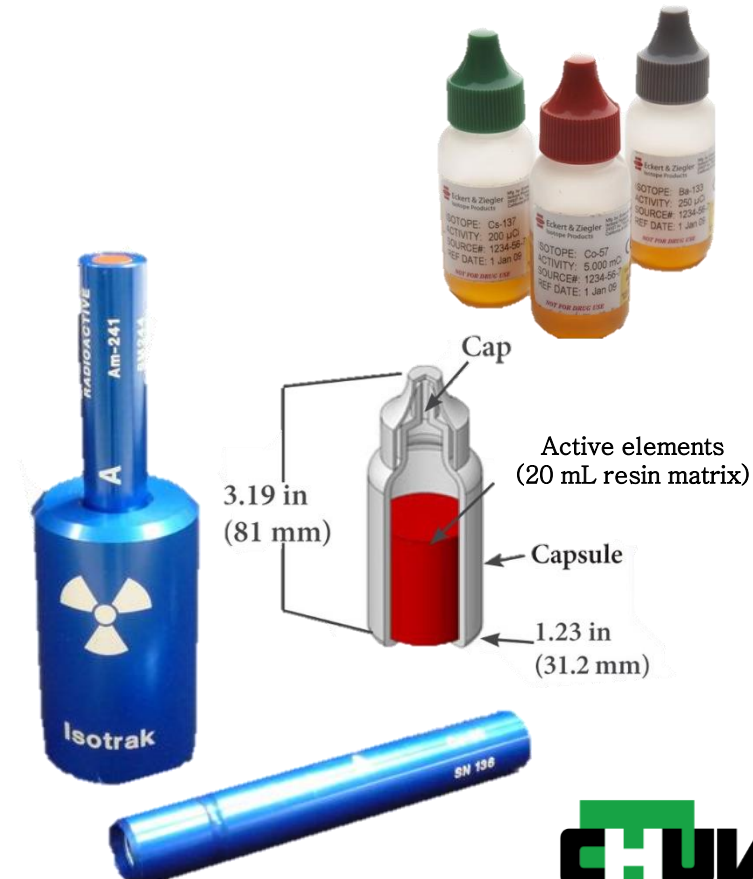
Active:

- Real-time reading.
- Used for optimization (in point-like interventions).
- Can give a real-time dose/ dose-rate information.
- User-defined alarms can be set.
- Do not generally satisfy the requirements set by the Swiss Ordinance on Dosimetry (linearity, energy response, dose range...).
- No background subtraction.
- The dose is not recorded.
- Example :
 - Solid state semiconductors



RP devices – QC (quality control)

- The functioning of each RP device – dose rate meter, surface contamination monitor, activimeter – has to be checked prior its use → control source (long-lived, e.g. Cs-137).
- Additionally, other QC may be required:
 - Dose rate meter / surface contamination monitor
 - Verification by accredited laboratory every 3 years
 - Activimeter checks
 - Daily (Cs-137 source, Cs-137 factor)
 - Weekly (Cs-137 source, measuring common factors)
 - 6 months: linearity
 - Every year : intercomparison / verification or calibration

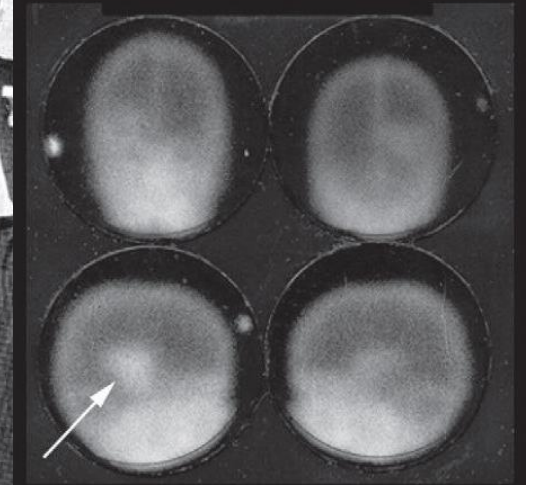
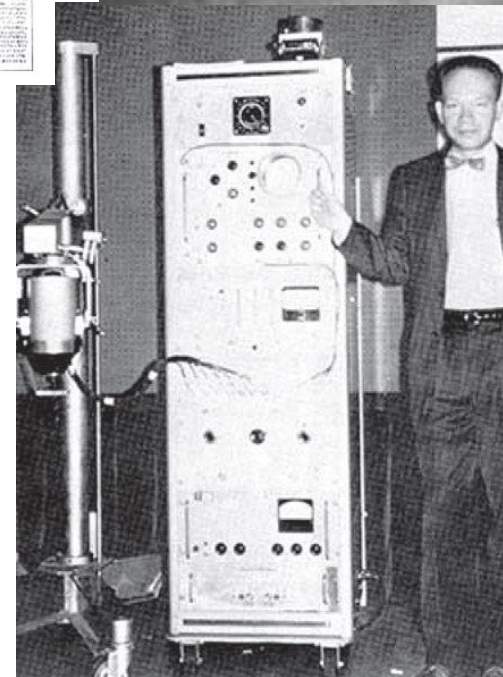
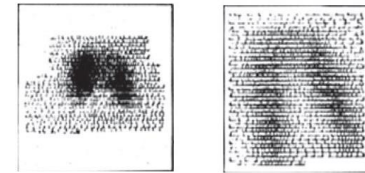
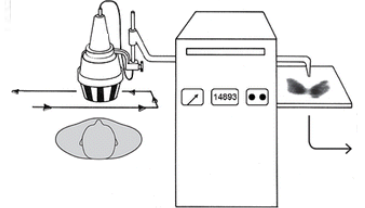
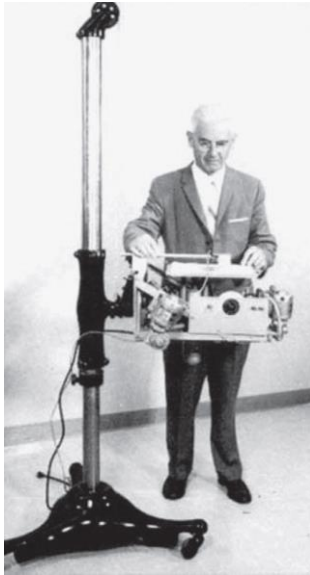


Part 2 – The gamma camera

Gamma camera & SPECT



Historical background



Rectilinear scanner by B. Cassen (1950)

Developed for planar images with I-131.

Scintillator counter scanning the area of interest.

Ink intensity \propto measured photon count-rate.

Long imaging times due to limited field of view.

First Gamma Camera by

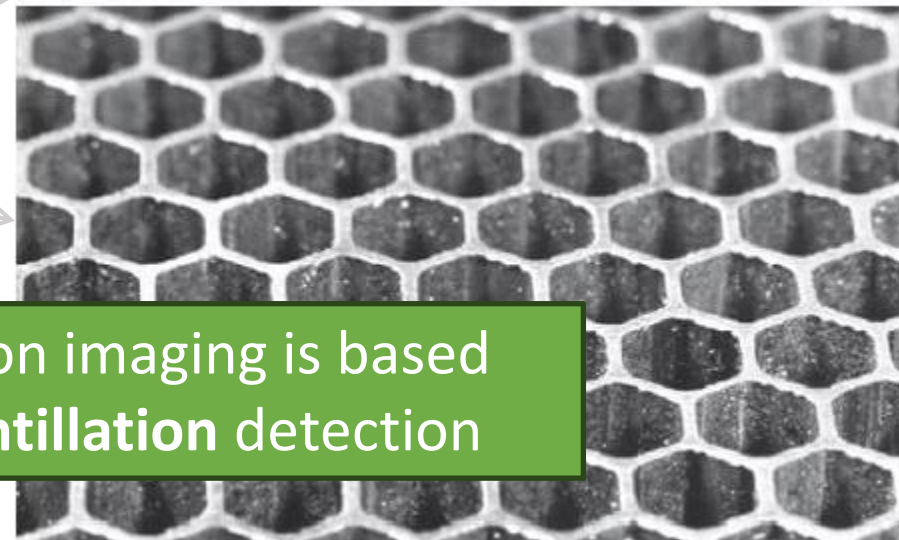
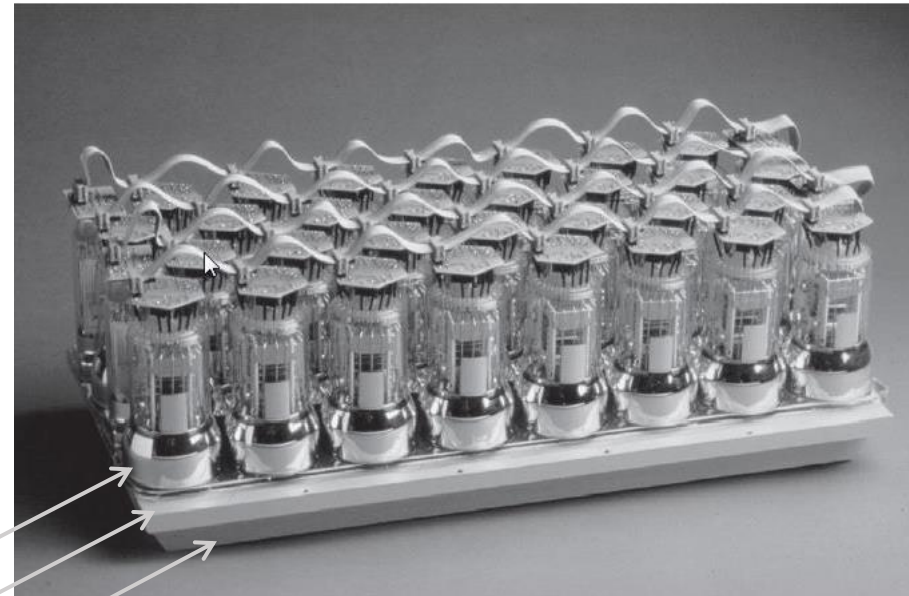
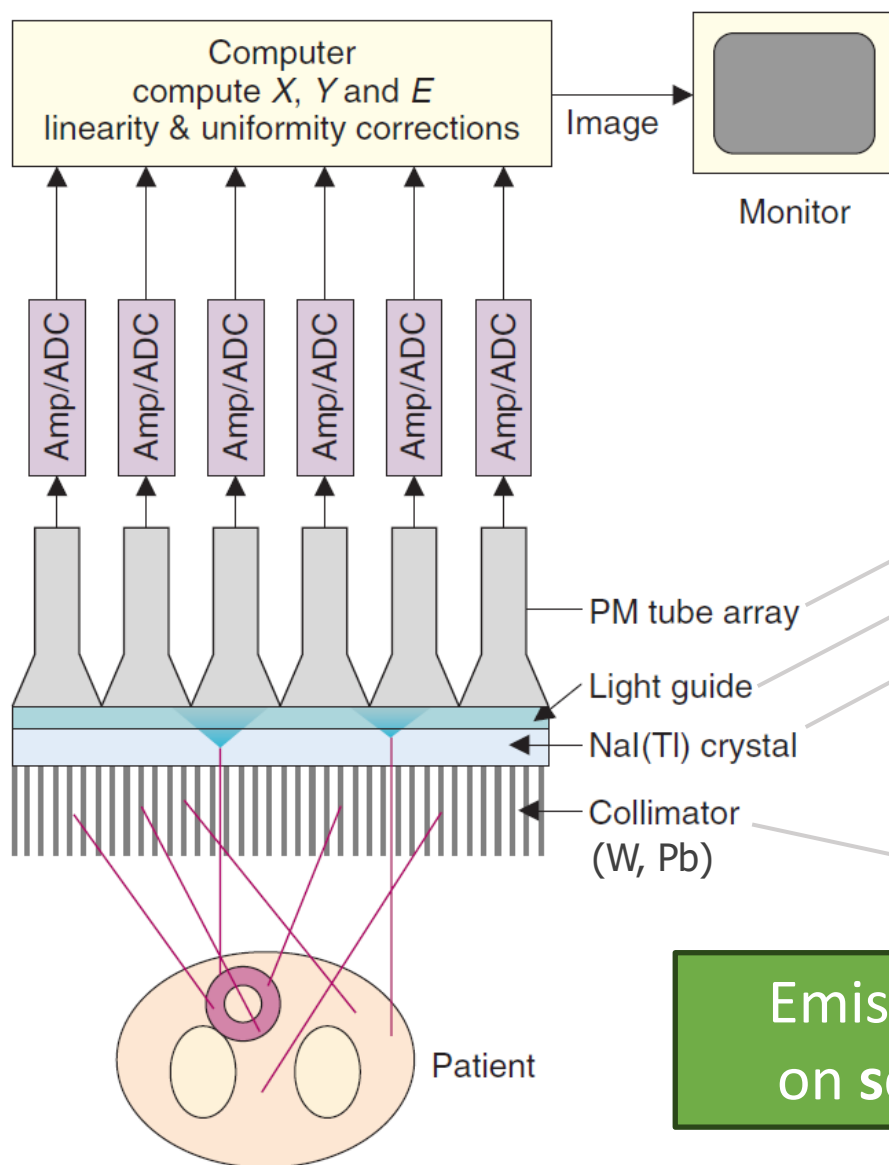
H. Anger (1958), using

Tc-99m Pertechnetate.

Brain scan of a patient with glioma.

Source: Physics in Nuclear Medicine, 4th edition, ISBN: 978-1416051985;
 CANCER BIOTHERAPY & RADIOPHARMACEUTICALS-Volume 15-
 Number 5-2000-Mary Ann Liebert, Inc. "Benedict Cassen: The Father
 ofBody Organ Imaging";
<https://doi.org/10.21019/9781582122830.ch6> ;
 Kemp HBS, Johns DL, McAlister J, Goodlee JN. The role of fl uorine-
 18 and strontium-87m scintigraphy in the management of infective
 spondylitis. J Bone Joint Surg. 1973;55:301-11. ;
https://en.wikipedia.org/wiki/Hal_Anger

Gamma camera: main components



Emission imaging is based on scintillation detection

Parallel hole collimator

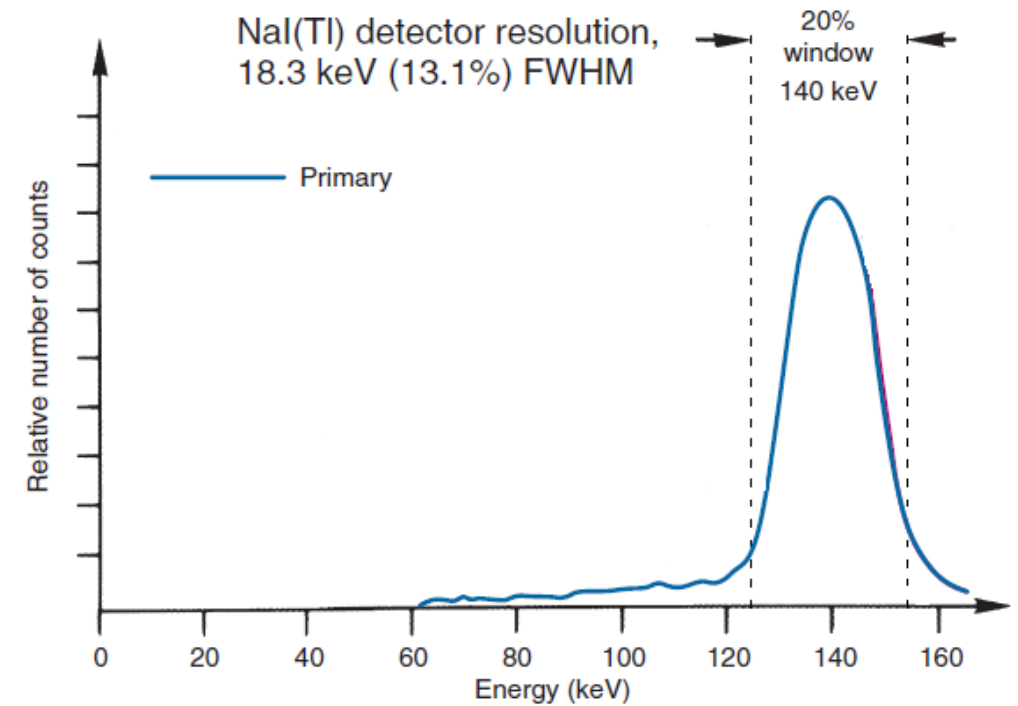
Recap : scintillation detectors

Scintillation material { Organic (liquids)
Inorganic (crystals: NaI(Tl), BGO, LSO, GSO, CsI, LaBr₃, plastic,...)

PROPERTIES OF SOME SCINTILLATOR MATERIALS

Property	NaI(Tl)	BGO	LSO(Ce)	GSO(Ce)
Density (g/cm ³)	3.67	7.13	7.40	6.71
Effective atomic number	50	73	66	59
Decay time (nsec)	230	300	40	60
Photon yield (per keV)	38	8	20-30	12-15
Index of refraction	1.85	2.15	1.82	1.85
Hygroscopic	Yes	No	No	No
Peak emission (nm)	415	480	420	430

^{99m}Tc $E_\gamma=140$ keV \rightarrow scintillation photon yield = $38 \cdot 140 = 5320$



Amount of scintillation light \propto deposited energy
 \rightarrow energy discrimination possible

Scintillation detectors – sodium iodide (NaI)

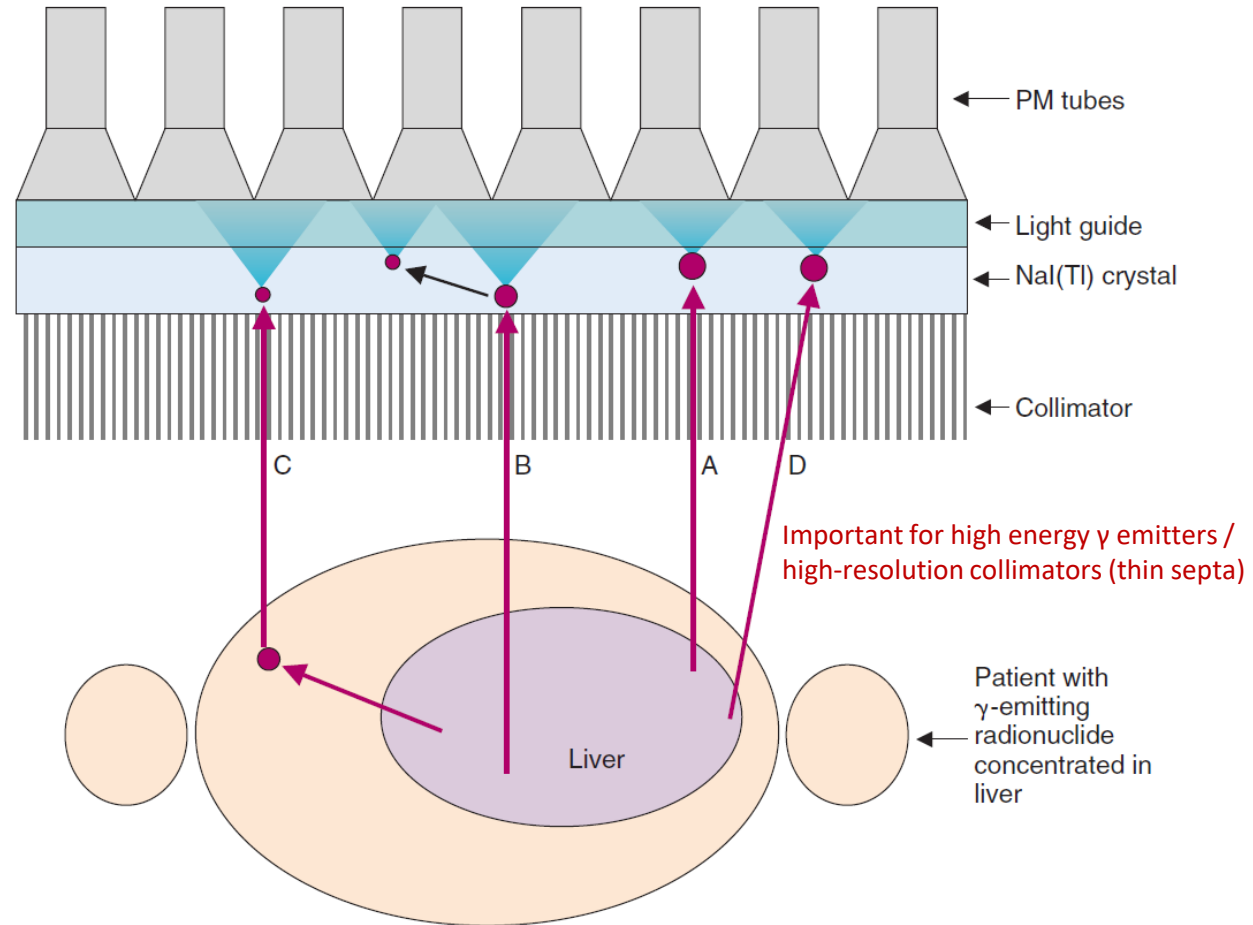
Advantages of NaI detectors:

- ✓ Reasonably dense and contains I (relatively high atomic number):
 - good absorber and efficient for penetrating radiations → ideal 50-250 keV photons.
 - predominant interaction in this range → photoelectric effect.
- ✓ Efficient scintillator (high photon yield).
- ✓ Transparent to its own scintillation (limited light loss due to self-absorption).
- ✓ Relatively inexpensive to grow in large plates.
- ✓ Scintillation light emitted by NaI is optimal (good wavelength) for PMT operation.

Disadvantages of NaI detectors:

- ✗ Crystal is fragile and easily fractured by mechanical or thermal stress (fractures = reduced amount of light reaching the photocathode).
- ✗ NaI is hygroscopic (humidity creates yellowish stains → impair light transmission)
- ✗ At energies > 250 keV, Compton effect is predominant.

Gamma camera: event localization



A) Valid event (useful for correct localization on the imaged region)

B) Scatter within the detector

C) Scatter within the patient

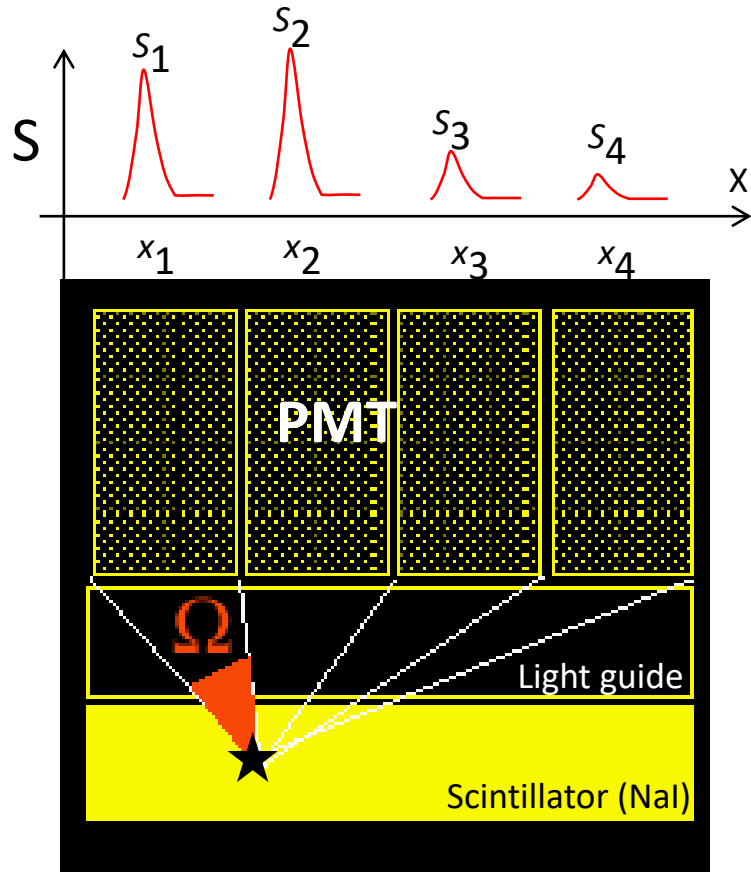
D) Septal penetration

Incorrect source localization !

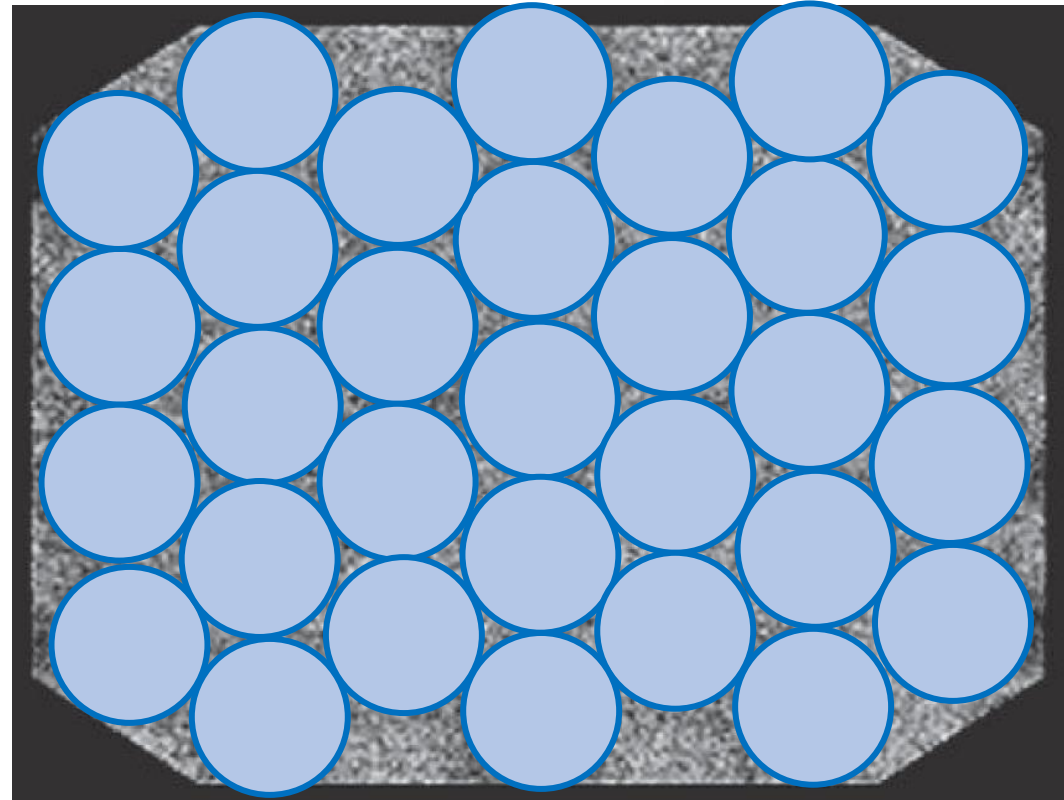
(+ loss of contrast, blurring)

Gamma camera: event localization

1 dimension



2D localisation on the crystal =
weighted average (centroid) of
the photomultiplier tube
response



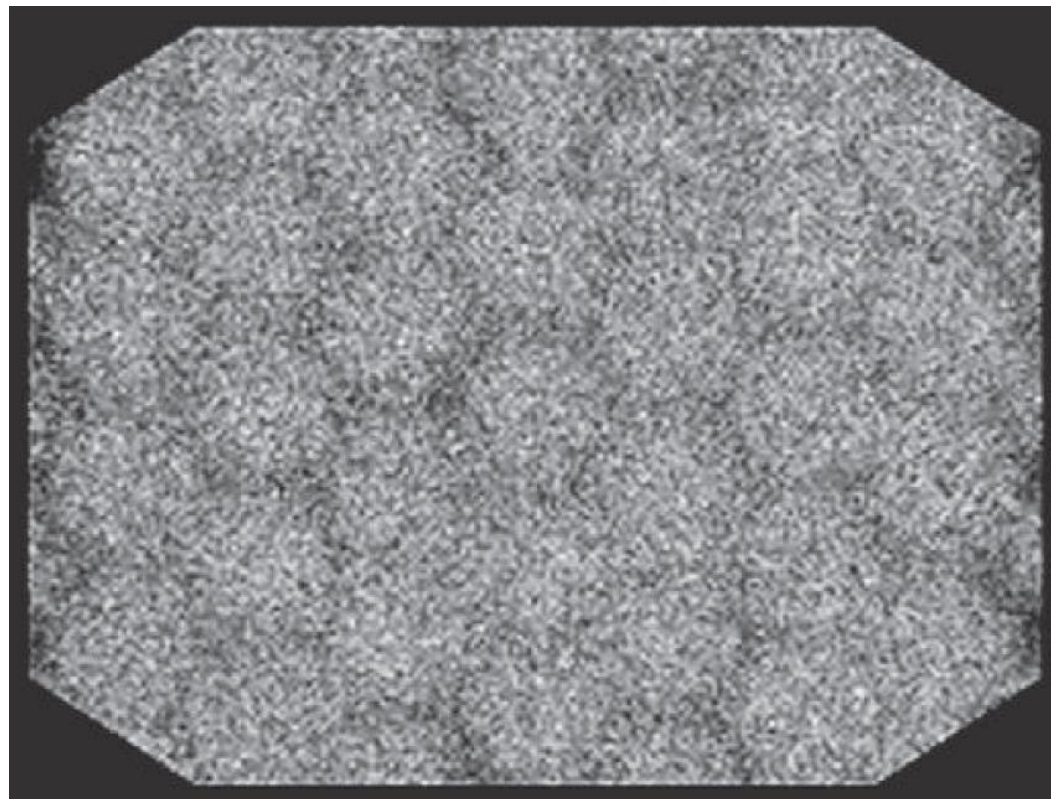
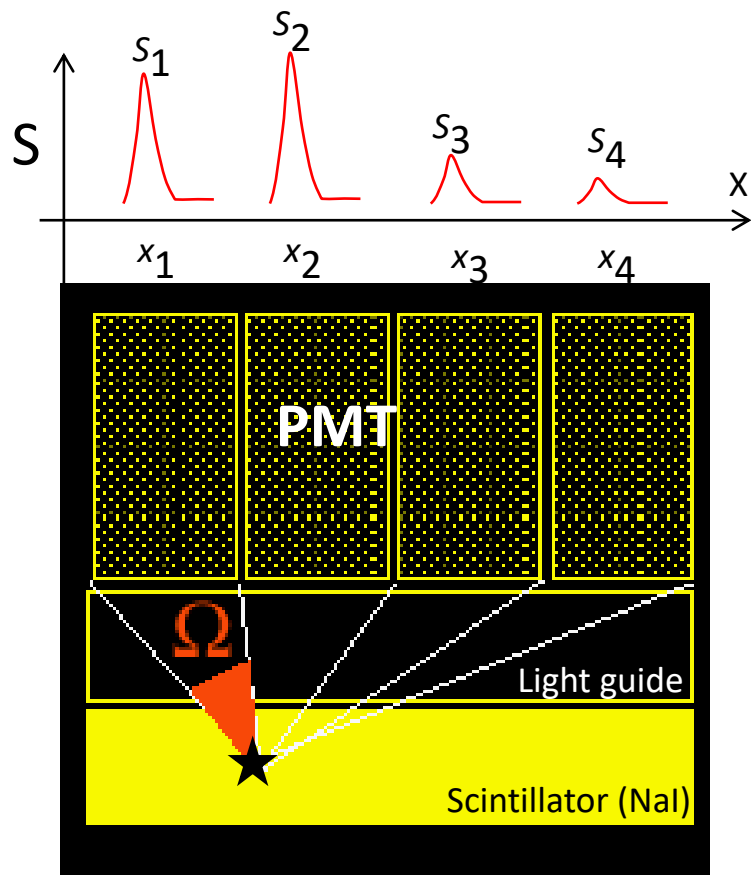
30-100 PMT

! PMT response does not vary linearly with the interaction position + other non-uniformities (crystal inhomogeneities, light reflections, pulse-height spectrum \neq among PMT ...).

In digital cameras, output signal of each PMT is digitalized and event position is calculated by a software \rightarrow take into account for non-linearity.

Gamma camera: event localization

1 dimension



30-100 PMT

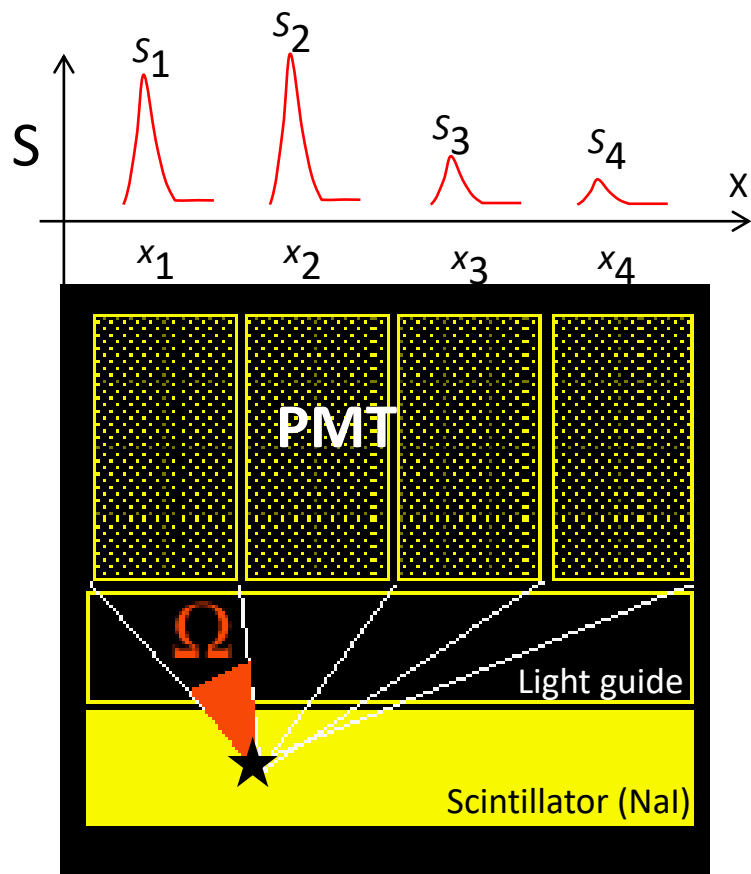
2D localisation on the crystal =
weighted average (centroid) of
the photomultiplier tube
response

⚠ PMT response does not vary linearly with the interaction position + other non-uniformities (crystal inhomogeneities, light reflections, pulse-height spectrum \neq among PMT ...).

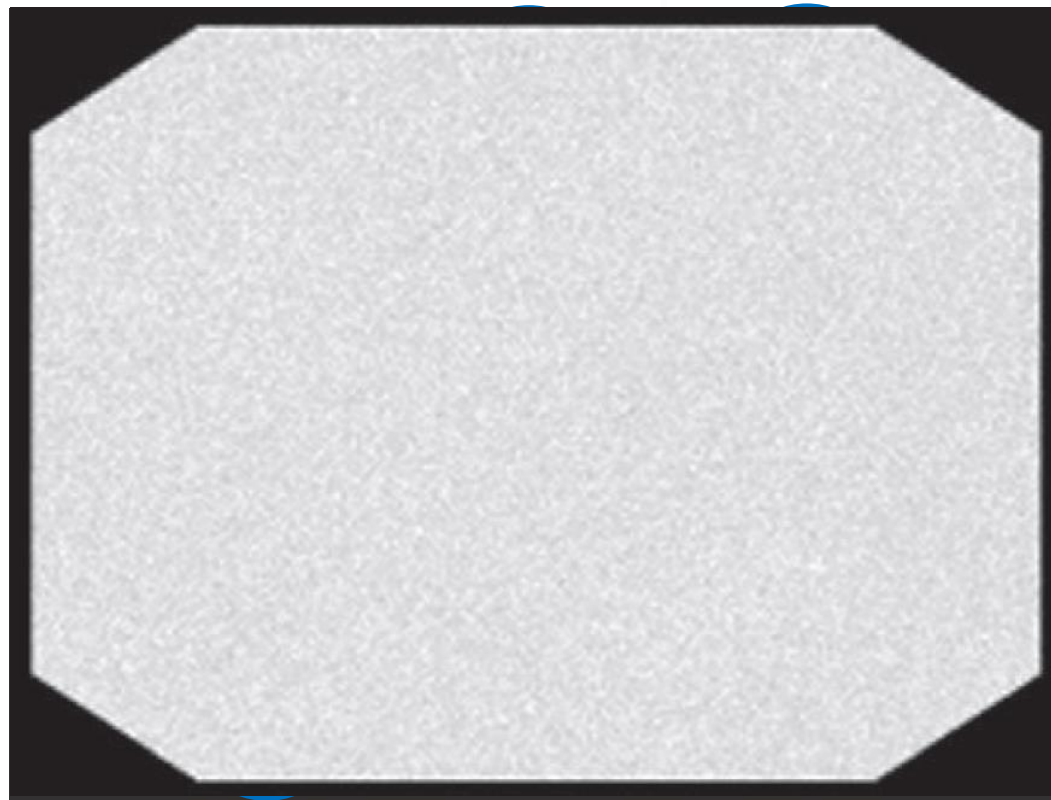
In digital cameras, output signal of each PMT is digitalized and event position is calculated by a software \rightarrow take into account for non-linearity.

Gamma camera: event localization

1 dimension



2D localisation on the crystal =
weighted average (centroid) of
the photomultiplier tube
response



30-100 PMT

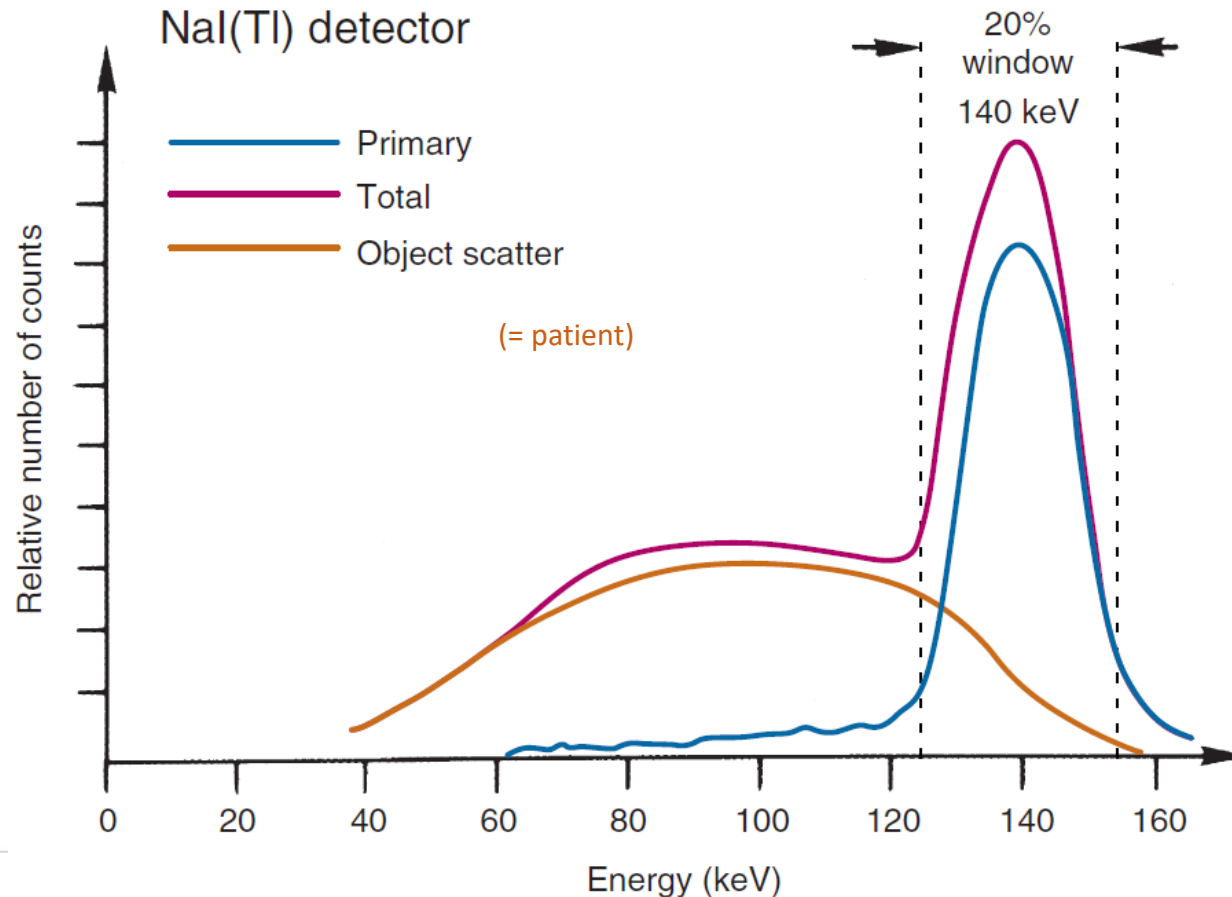
⚠ PMT response does not vary linearly with the interaction position + other non-uniformities (crystal inhomogeneities, light reflections, pulse-height spectrum \neq among PMT ...).
In digital cameras, output signal of each PMT is digitalized and event position is calculated by a software \rightarrow take into account for non-linearity.

Gamma camera: energy determination

Why is energy discrimination so important ?

Scattered photons are undesirable : no useful information of the source distribution!

They also lose energy along their path → can be discriminated from using the pulse-height analyser (PHA) present on each PMT (signal amplitude \propto scintillation light \propto deposited energy in the crystal).



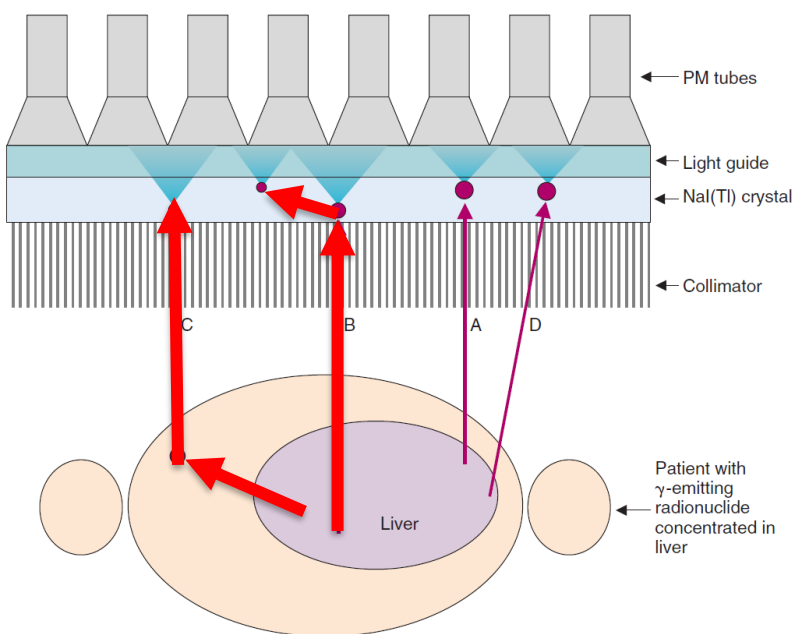
PE effect :

they mostly result in full deposition of the photons in the detector (photopeak).

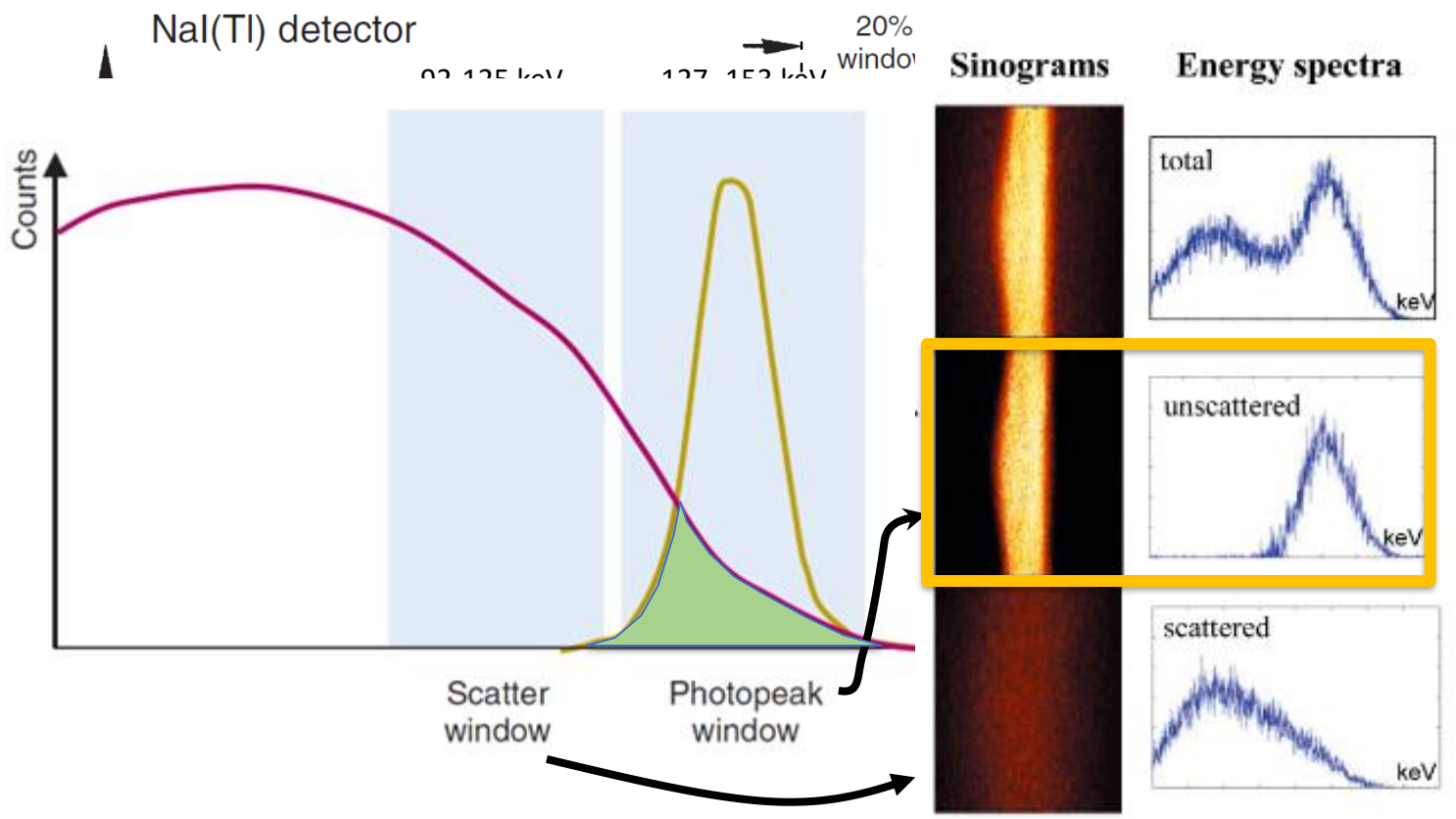
Compton :

part of the energy is transferred to the detector via the recoil e^- , the scattered photon may either be detected or escape the detector.
→ Only PE or Compton scattered at small angles should be accepted!

Gamma camera: energy determination → scatter correction



Diffusion, reduced contrast, overestimation of activity.

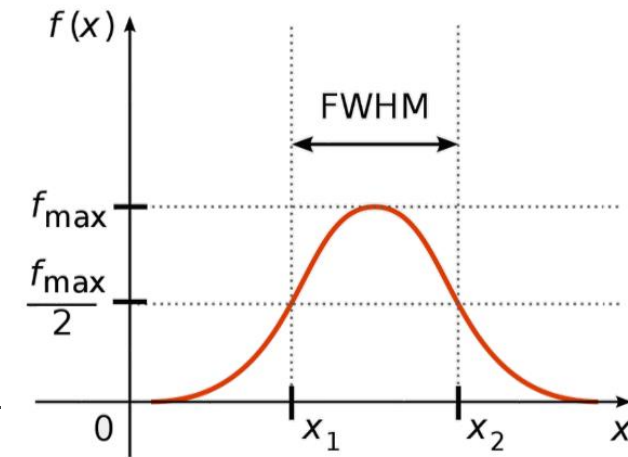
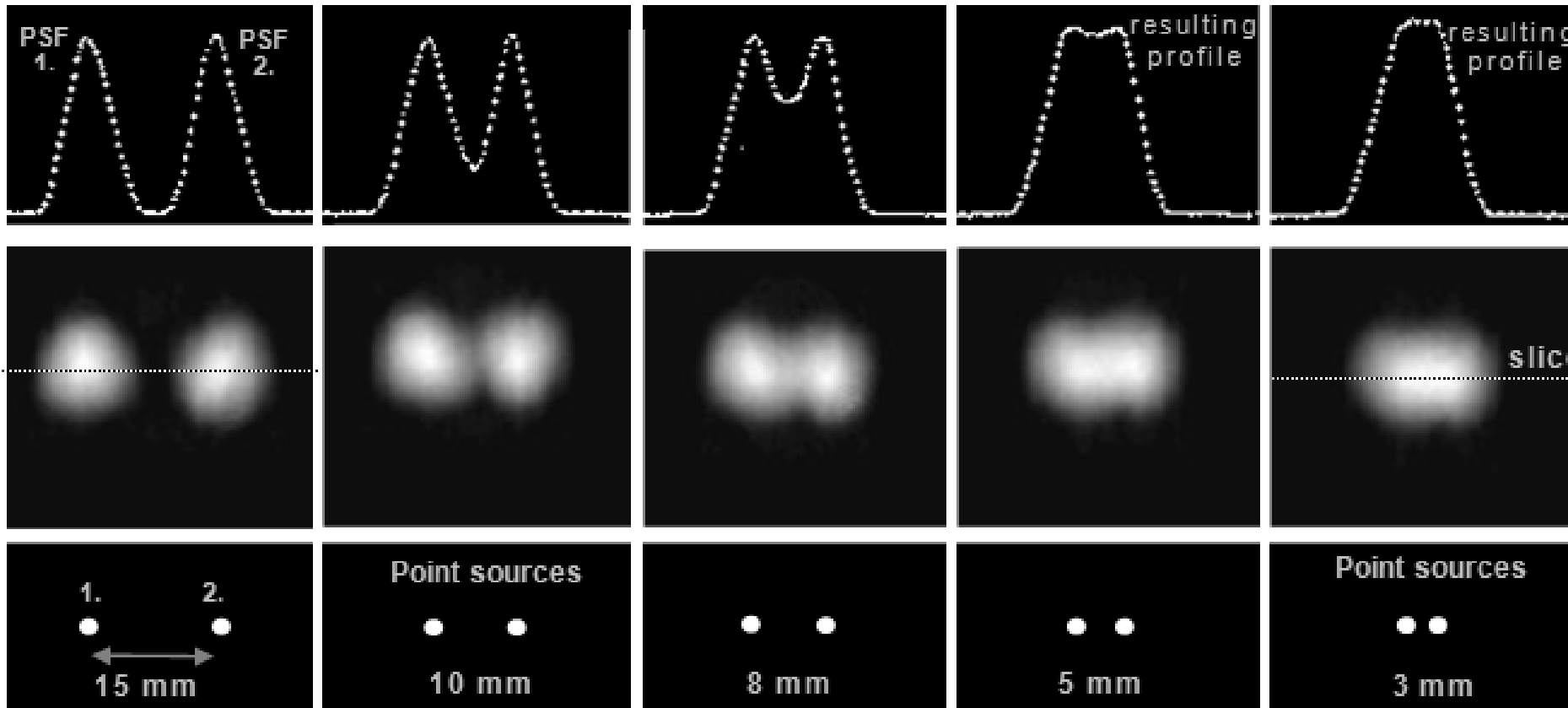


- Two (or more) energy windows: one for scattering and one for the photoelectric peak.
- The goal: to detect only those photons that have not scattered (= deviated from the path).
- For each projection, subtraction of the scattered events that fall into the window of the photopeak.

Spatial resolution

What defines the spatial resolution of an imaging device?

Ability to distinguish two adjacent structures as separate (sharpness/detail).



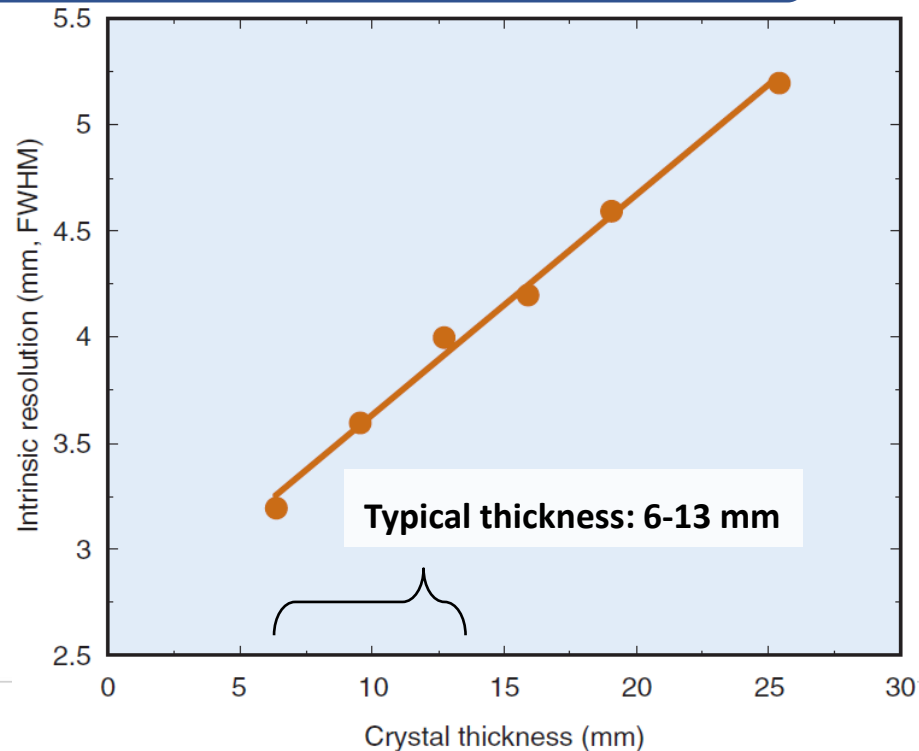
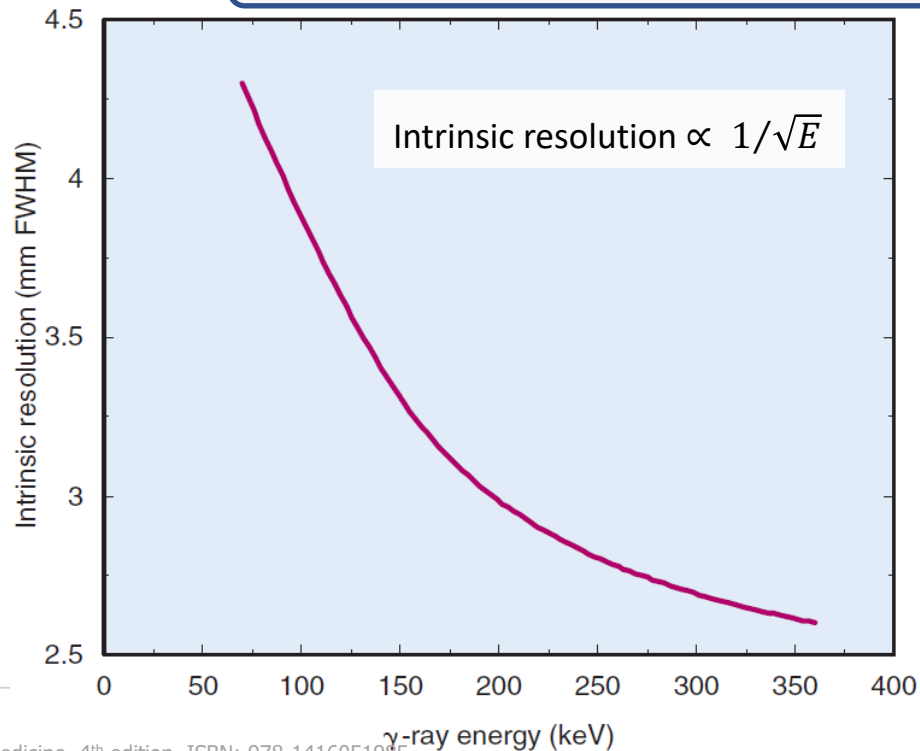
Small FWHM =
Better resolution!

Spatial resolution

Intrinsic spatial resolution: achievable by the detector itself and electronics (no collimator). Depends on:

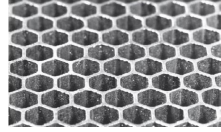
- **γ -ray energy** : low energy γ produce less scintillation photons per event \rightarrow statistical fluctuations.
- **crystal thickness** : thicker detectors are more efficient in detecting radiation 👍 (spoiler), but:
 - Increased light spreading before reaching PMT 👎
 - Greater probability of detecting multiple Compton-scattered events 👎

For modern cameras: $\sim 3 - 4.5$ mm (of FWHM for Tc-99m)



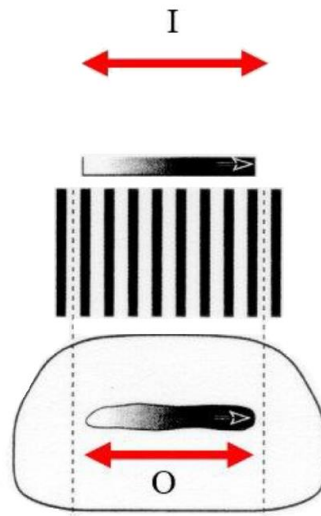
Spatial resolution

Extrinsic spatial resolution : also influenced by linear and angular sampling, reconstruction algorithm (incl. spatial smoothing) and mostly by the *presence of collimators*.



Collimators :

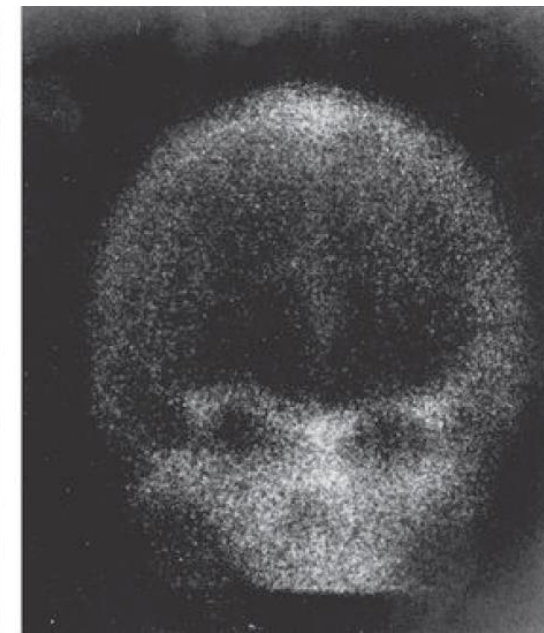
- 👍 improve source distribution localisation by selectively allowing only photons travelling in specific directions to hit the detector (absorptive collimation)
- 👎 degrade spatial resolution (weak link for the camera performance!).



Parallel-hole
Image of the same
size and orientation
(most common)



Parallel-hole collimator



Converging collimator

Spatial resolution

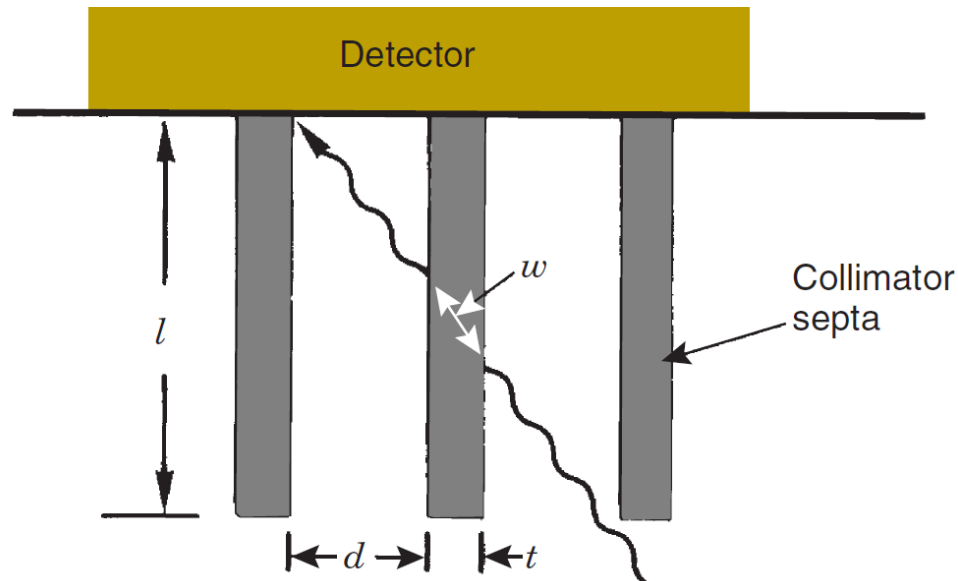
Collimator **efficiency** : fraction of γ striking the collimator that reach the detector.

Collimator **resolution**: sharpness of detail of the object projected on the detector.

Both influenced by **(1) septal penetration** and **(2) geometry of the collimator holes**:

1) Septal penetration should be reasonably small ($< 5\%$) \rightarrow thinner septa as possible (better resolution and efficiency) using highly absorbing material (Pb, W, Ta).

⚠ Septa thickness depends on γ energy
 \rightarrow low- (< 140 keV), medium- (~ 300 keV) & high-energy collimators (> 300 keV) available!

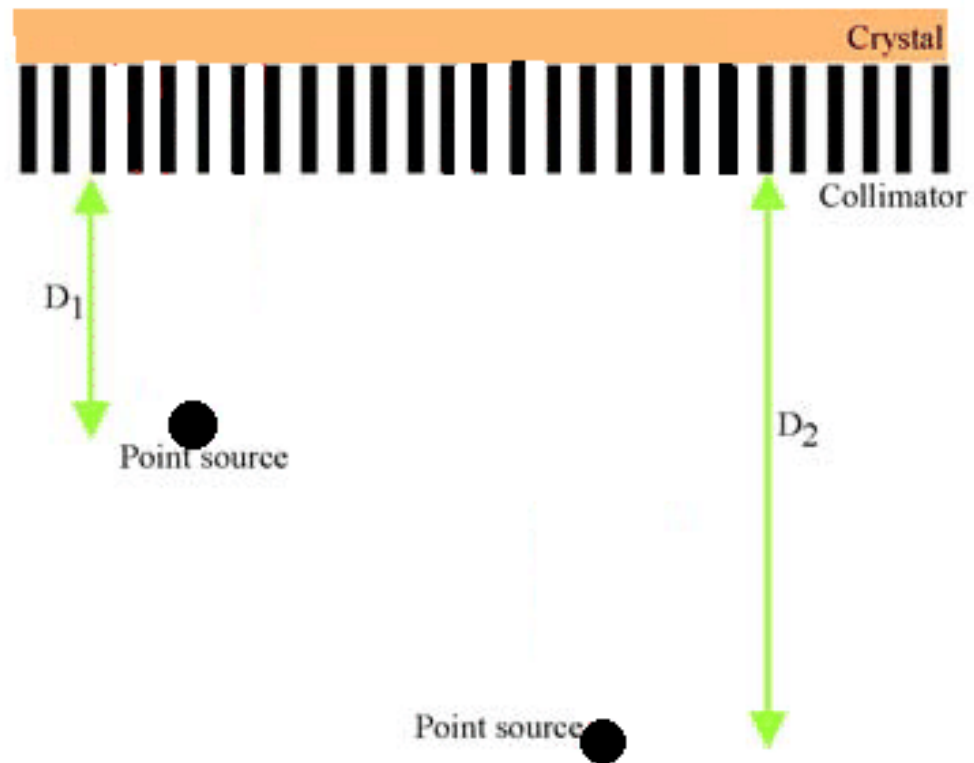


^{131}I Effect on Different Collimation ($\gamma = 364$ keV)



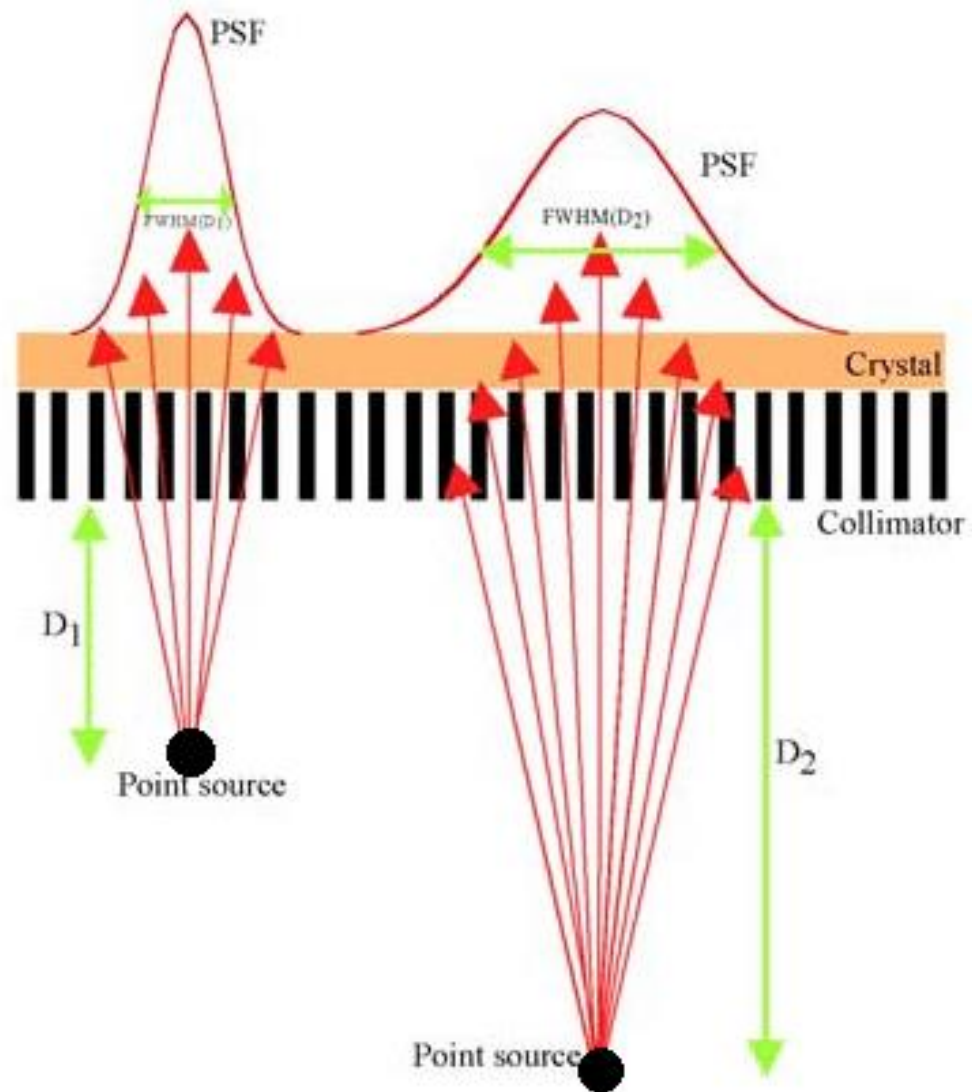
Spatial resolution

Quiz: which configuration provides a better spatial resolution?



Spatial resolution

Quiz: which configuration provides a better spatial resolution?



Spatial resolution

Collimator **efficiency** : fraction of γ striking the collimator that reach the detector.

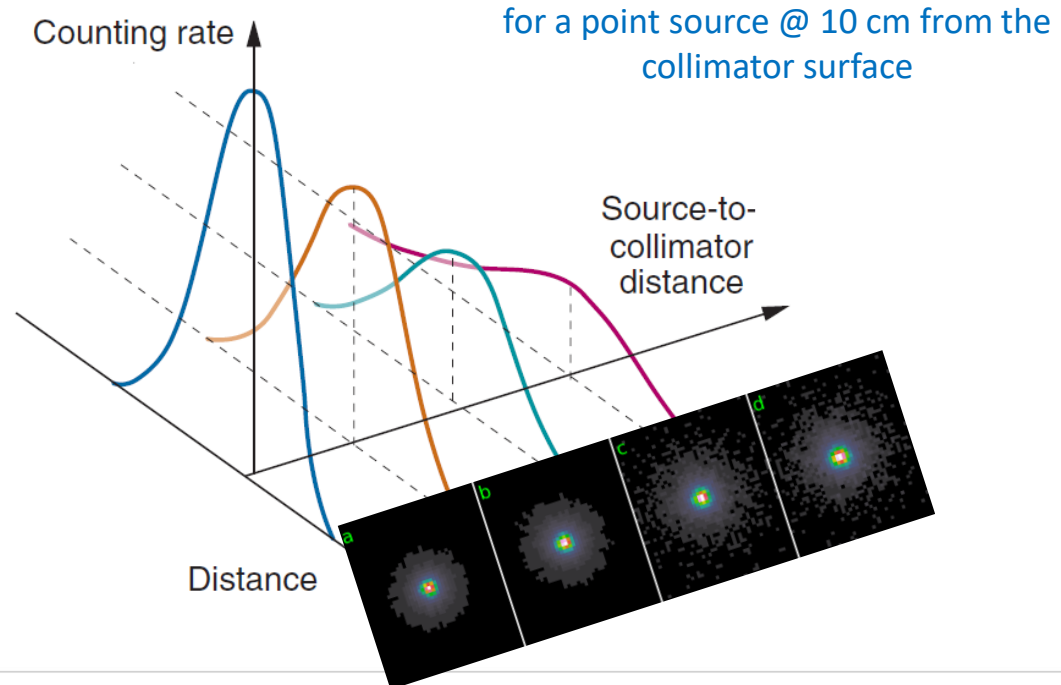
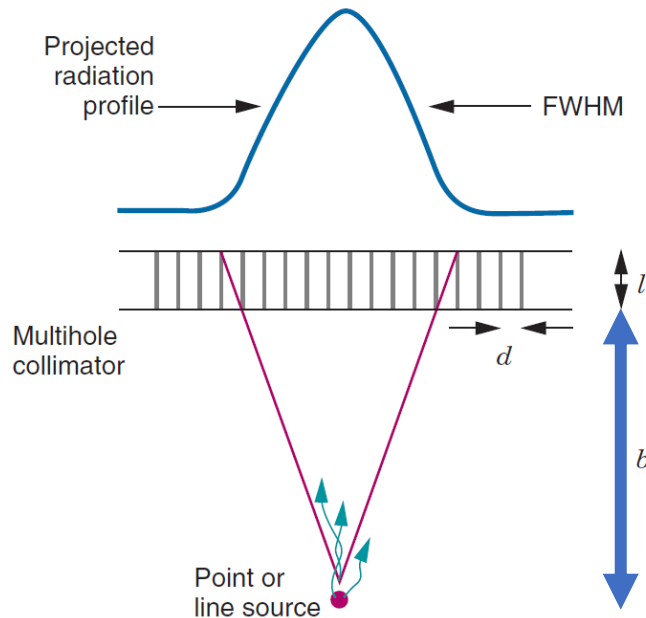
Collimator **resolution**: sharpness of detail of the object projected on the detector.

Both influenced by (1) septal penetration and (2) **geometry** of the collimator holes:

2) Geometry (shape, length, diameter) and distance from the source.

$$R_{coll} \approx d(l + b) / l$$

Typical $R_{coll} \sim 10 \text{ mm}$ (7 – 13 mm)



*In equation, l should be the effective length l_{eff} of the collimator holes (slightly smaller than the physical one to take into account the effect of septal penetration, especially higher-energies collimators). Here it is referred as « l » for the sake of simplicity.

Spatial resolution

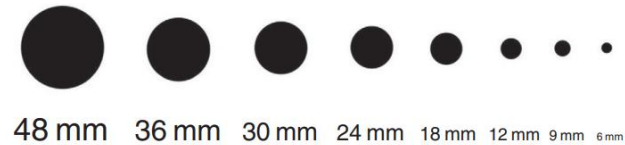
The gamma camera **total resolution** (or **system** resolution) is mainly influenced by the **intrinsic** resolution (detector and electronics) **and** **collimator** resolution.

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

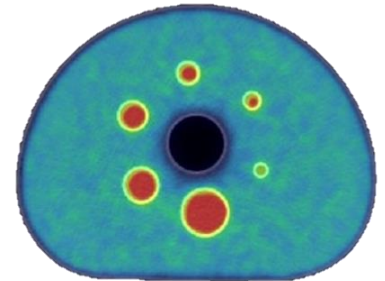
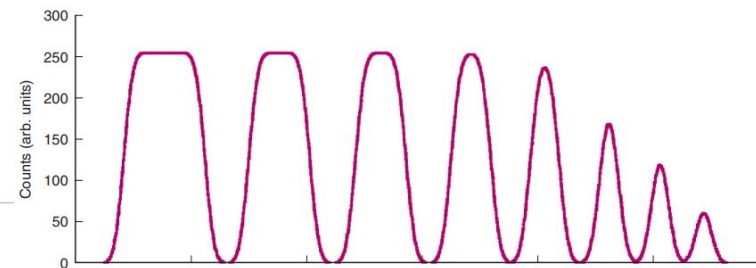
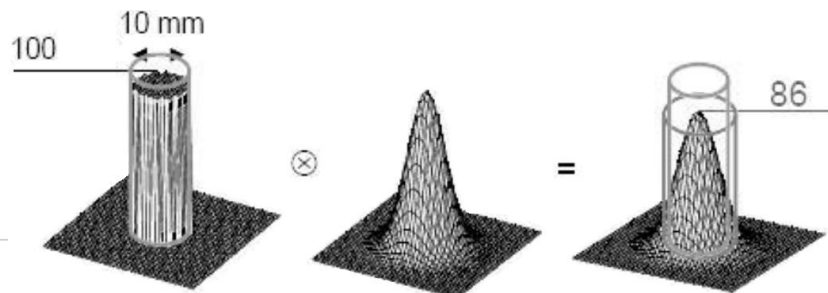
- It provides a metric of the “real” (clinically-significant) resolution.
- It depends on the source-to-collimator distance (R_{coll}).
- For distances of 5-10 cm (typical organ depth), it is governed by R_{coll} .
- It is degraded by scattered radiation.

Q: What is the system resolution of a typical gamma camera ?

$$R_{sys} = \sqrt{(3.5 \text{ mm})^2 + (10 \text{ mm})^2} = 10.6 \text{ mm}$$



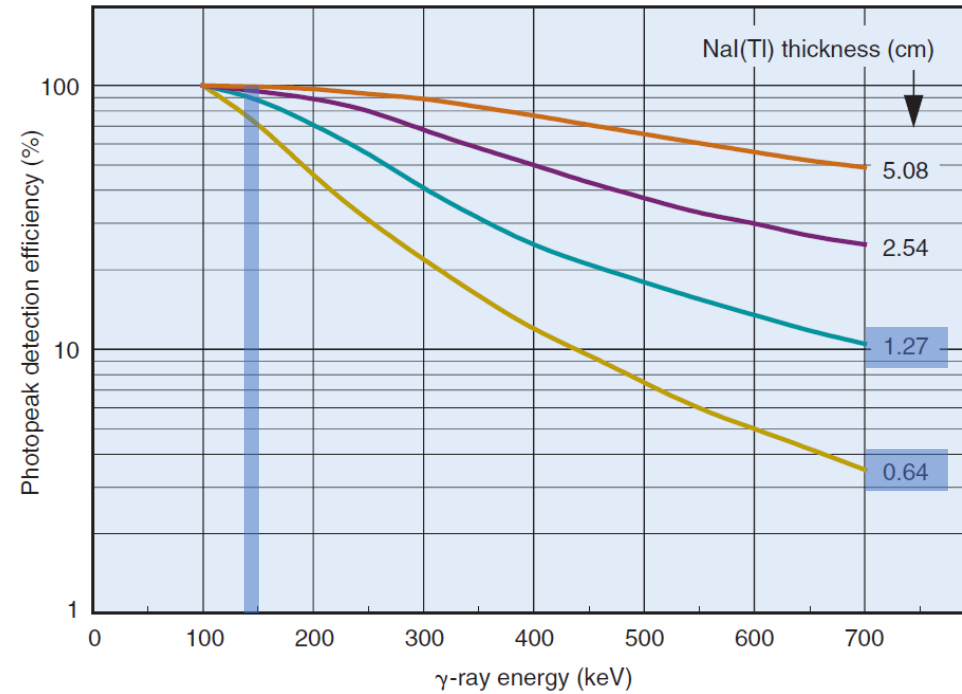
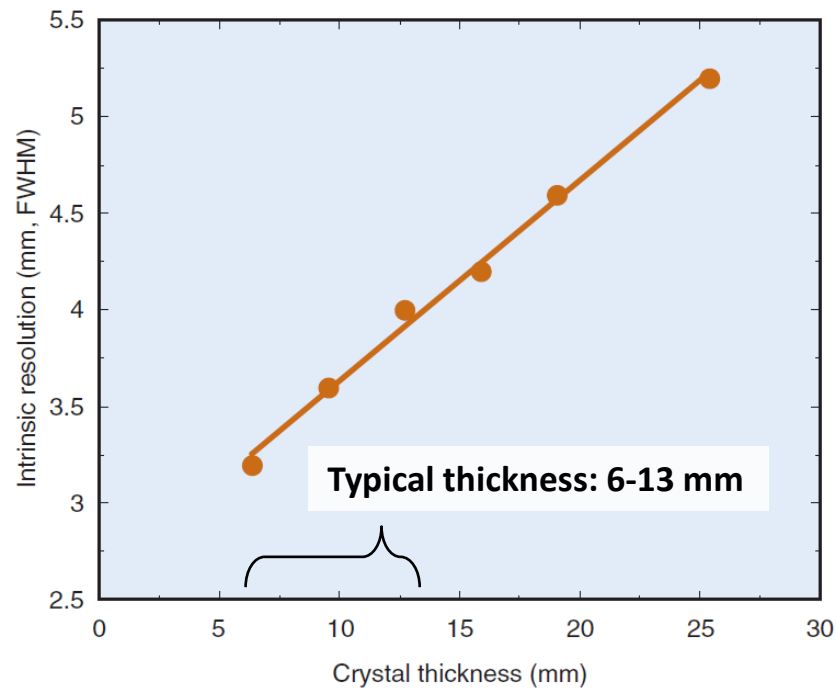
Finite resolution is partly responsible for partial volume effects :



Efficiency

Efficiency: ability to detect incoming radiation. Influenced by **detector** and collimator.

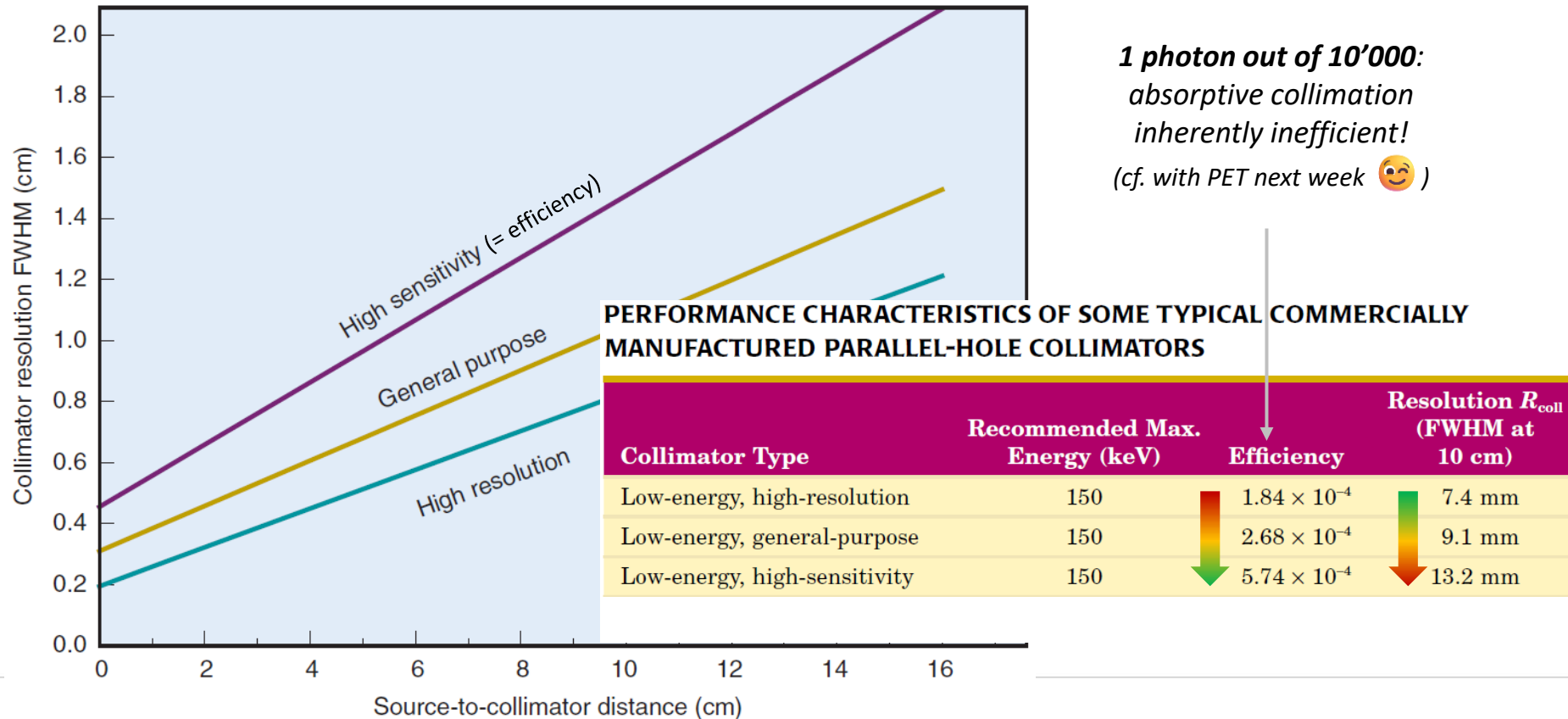
- Thicker detectors have higher intrinsic efficiency (more volume allowing the radiation to deposit energy within the crystal) BUT loss in spatial resolution.
- **Tradeoff:** acceptable detection efficiency and high intrinsic spatial resolution in the energy range of 100 - 200 keV.



Efficiency

Efficiency: ability to detect incoming radiation. Influenced by detector and **collimator**.

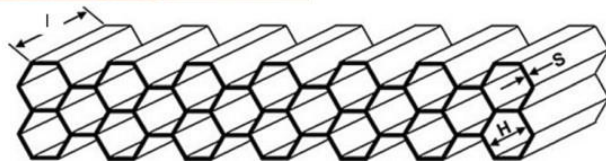
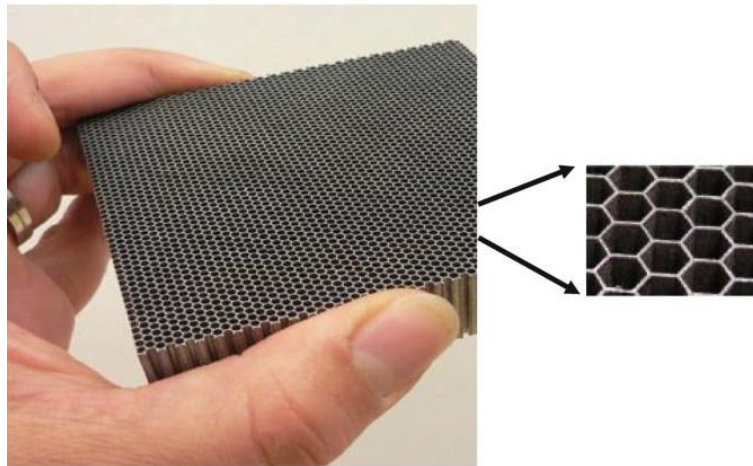
- Fraction of γ rays striking the collimator that pass through it.
- Efficiency (as resolution) is also influenced by the collimator geometry.
- Low-energy collimators have thinner septa and thus increased efficiency.
- **Tradeoff:** high efficiency results in degraded spatial resolution.



Spatial resolution & efficiency

The properties of the gamma camera may be either **intrinsic** (due to the detector) or **extrinsic** (due to the influence of the collimators).

*The **collimators are necessary** as they allow for the **correct localisation** of the source distribution within the patient, **but degrade** overall system **efficiency**.*



efficiency

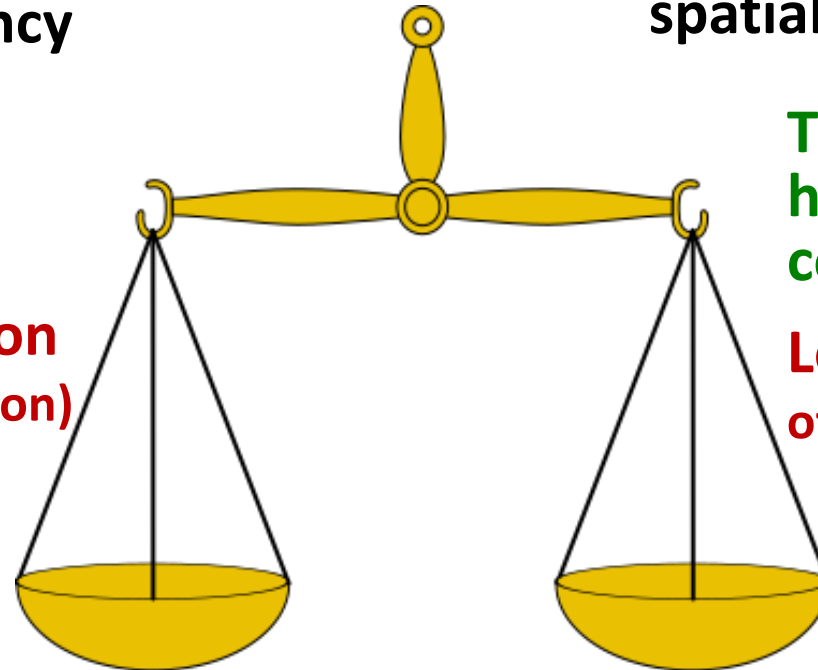
**Thick crystal
without
collimator**

**Loss of localisation
(and spatial resolution)**

spatial resolution

**Thin crystal,
highly selective
collimator**

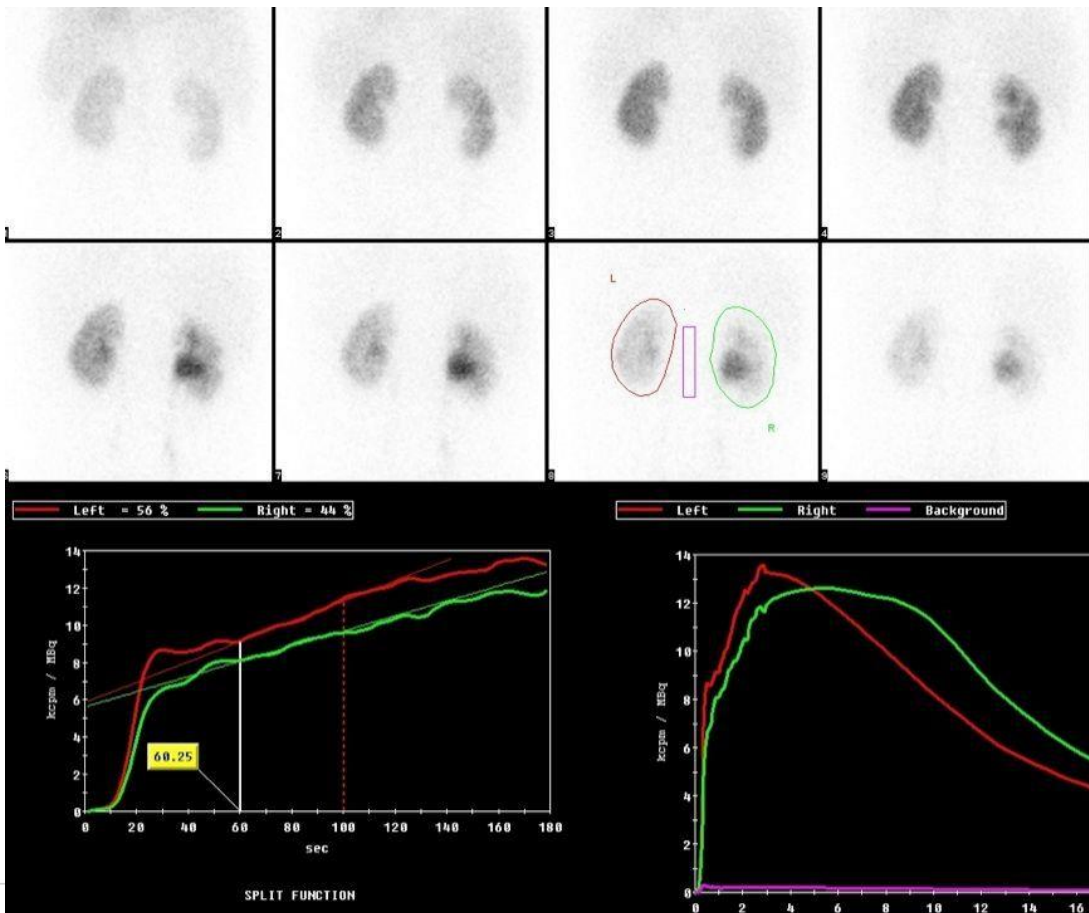
**Loss of signal (loss
of efficiency)**



Part 3 – SPECT

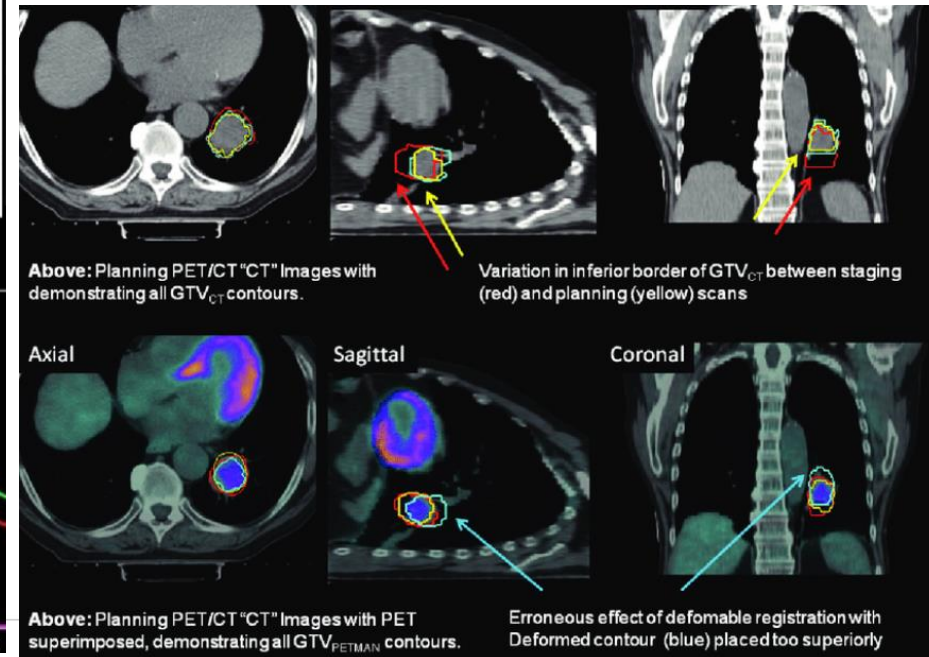
From planar scintigraphy to SPECT

- Use the gamma camera detection principle to image activity distribution within the patient.
- **Planar images** can be used for both:
 - **static** studies (activity distribution at a given given time).
 - **dynamic** studies (evolution over time) → e.g. kidney function.



Limitations:

- Signal from overlapping regions !
- No 3D information !



Introduction to tomography in nuclear medicine

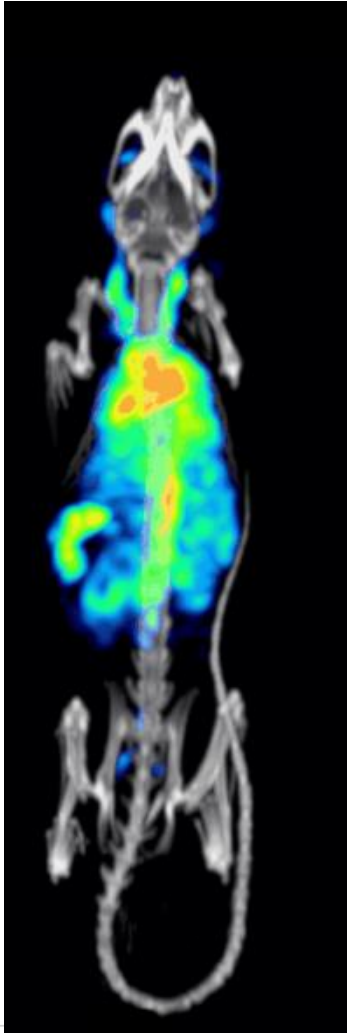
CT

TCT: **transmission**
computed tomography

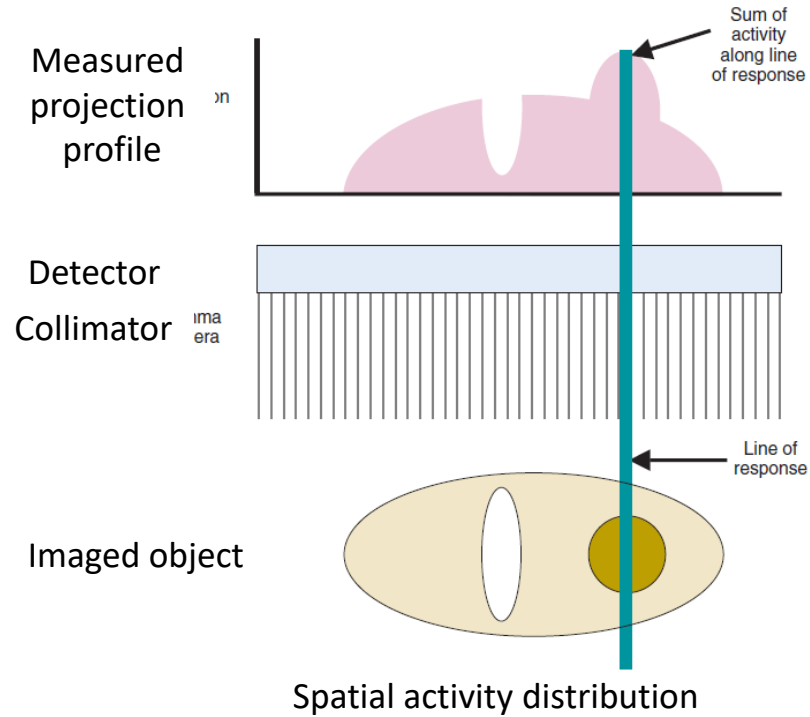
≠

SPECT / PET

ECT: **emission**
computed tomography



Introduction to tomography in nuclear medicine



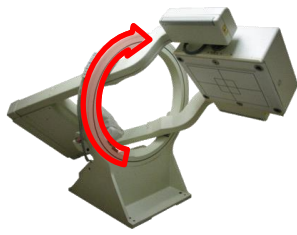
Projections in a simple 1-D detector case

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

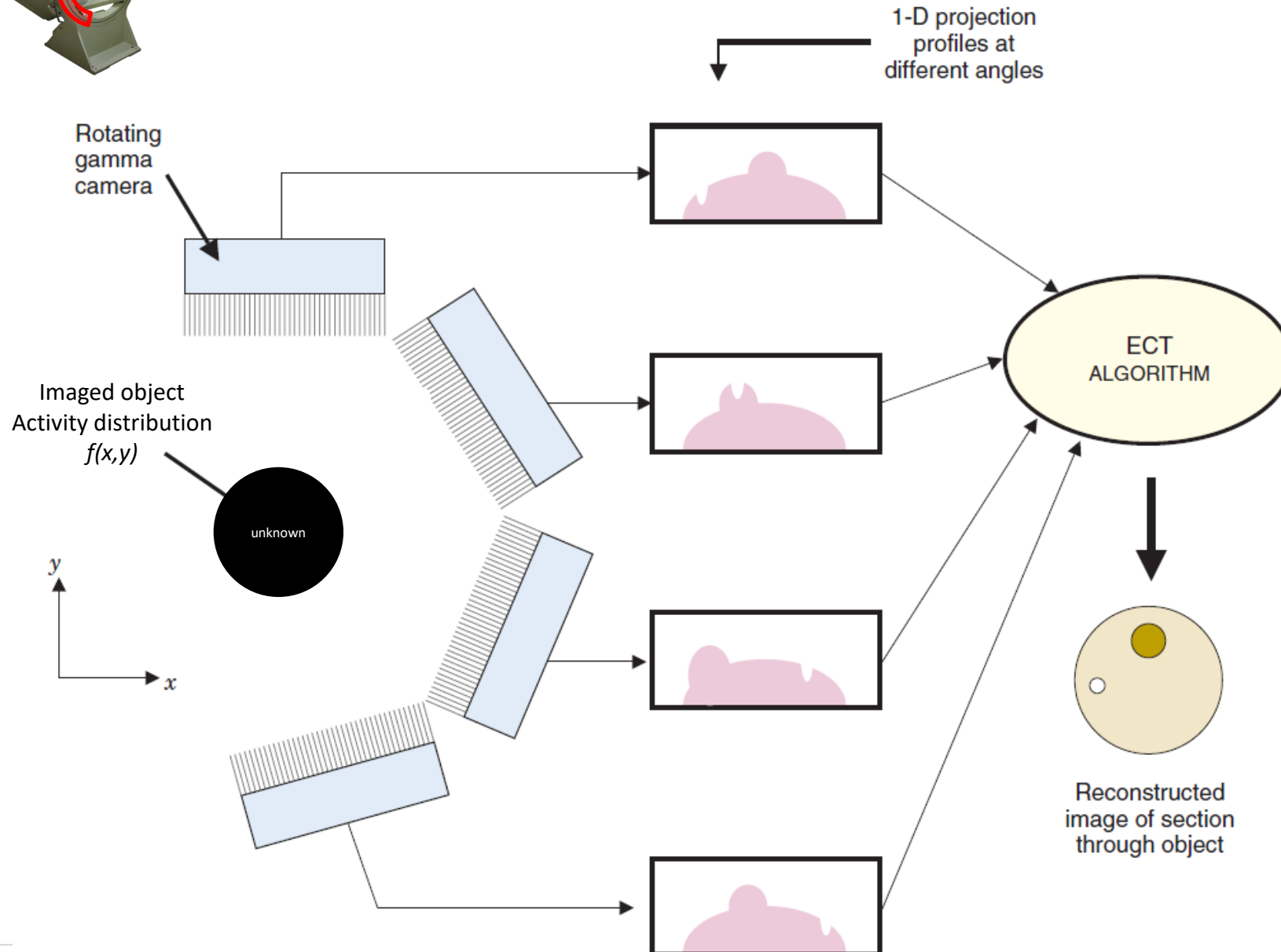


Head rotation around the imaged object

Many projections at different angles are obtained

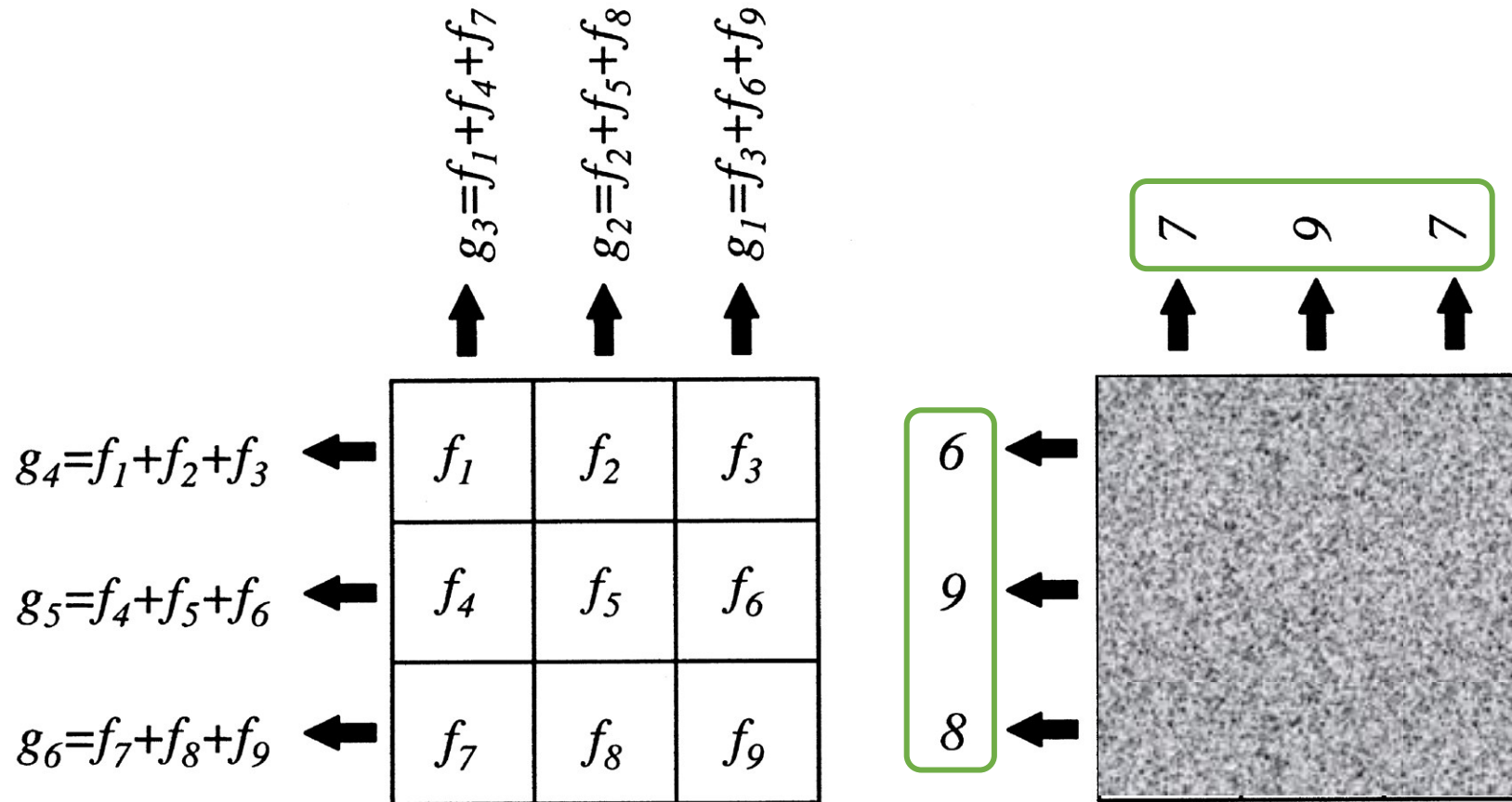


1-D projections of a unknown activity distribution $f(x,y)$

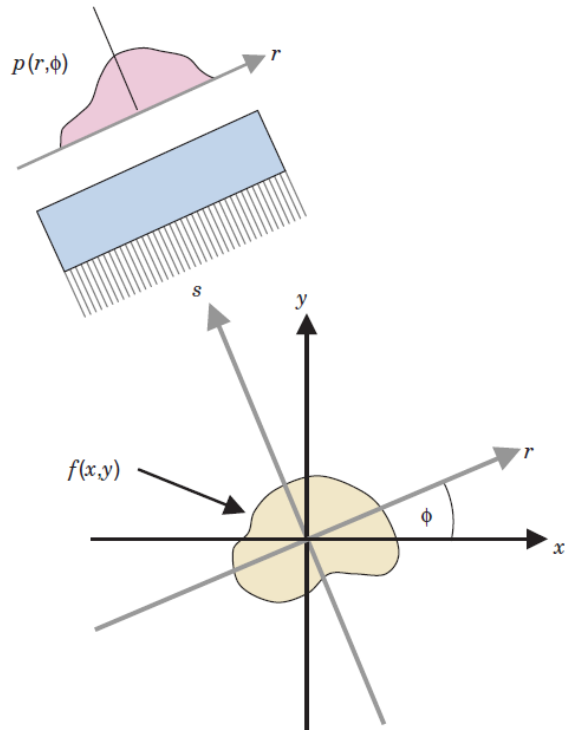
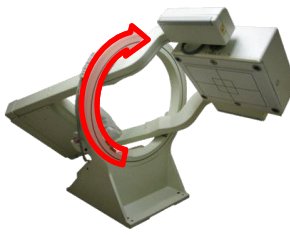


Forward projection

Principle of (forward) projection for one 3x3 slice at angle $\phi = 0$ and $\phi = 90^\circ$.



Obtain an image starting from projections



N projections are recorded around the emitting object (0-360 deg)

N projections $p(r,\phi)$

Angular spacing is $A_s = 360/N$ deg

Typically $A_s = 3$ deg for $N = 120$ projections

Dual-head camera \rightarrow 60 acquisition steps

15 sec per step \rightarrow 15 minutes image acquisition

Projection data p are arranged in (r, ϕ) coordinate system: tilted of angle ϕ wrt (x,y) and fixed wrt gamma camera

$(x,y) \rightarrow (r,\phi)$

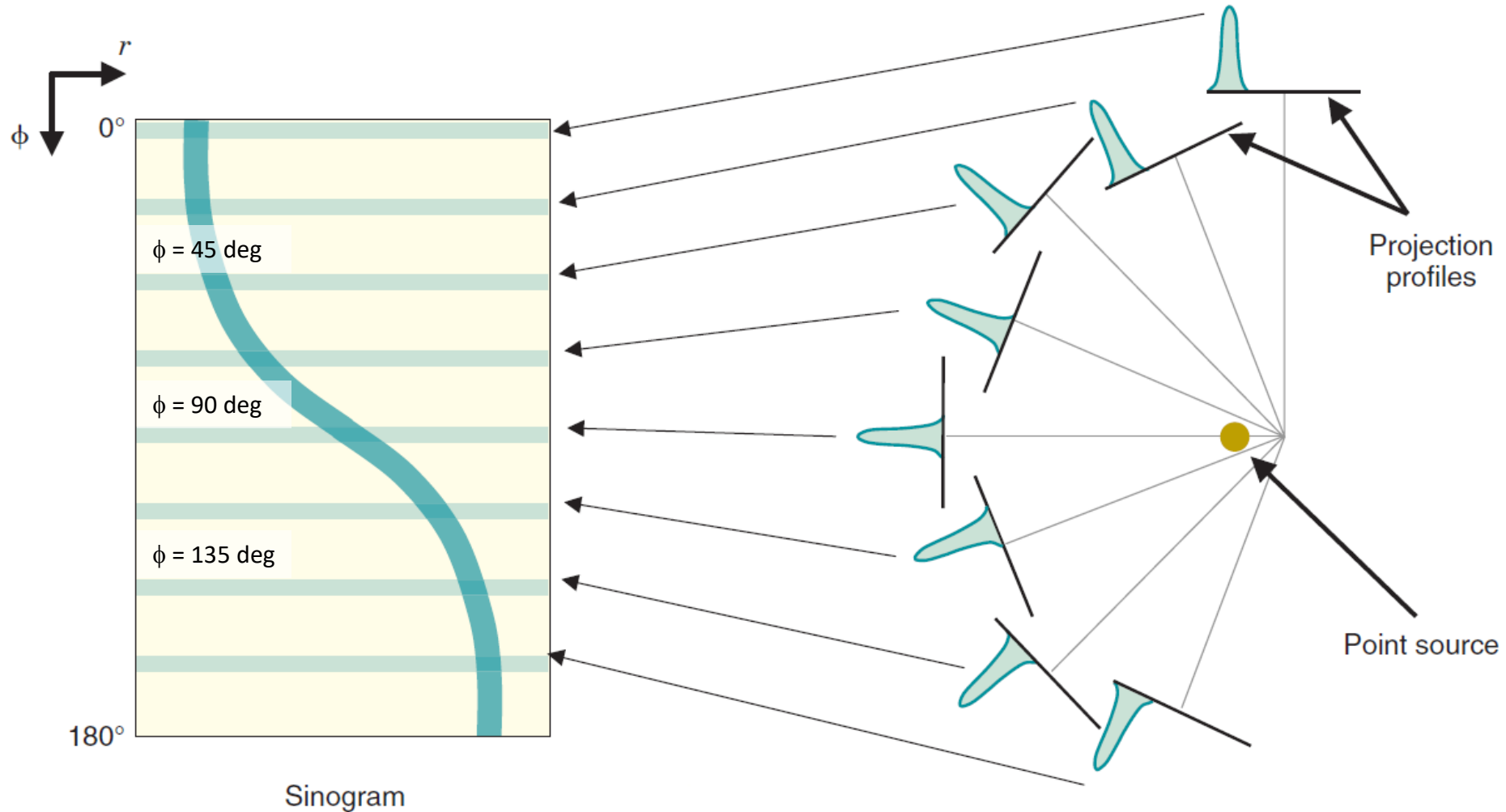
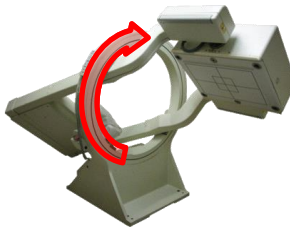
base for **sinogram** representation:
set of projection data in the form of a 2D matrix $p(r,\phi)$



dual-head camera = 2 projections per step

1D Projection data are arranged in a 2D (r, ϕ) **sinogram representation**

Each row (ϕ) in the sinogram displays the intensity profile measured in the corresponding projection



Each **row** (ϕ)

Each **column** (r)

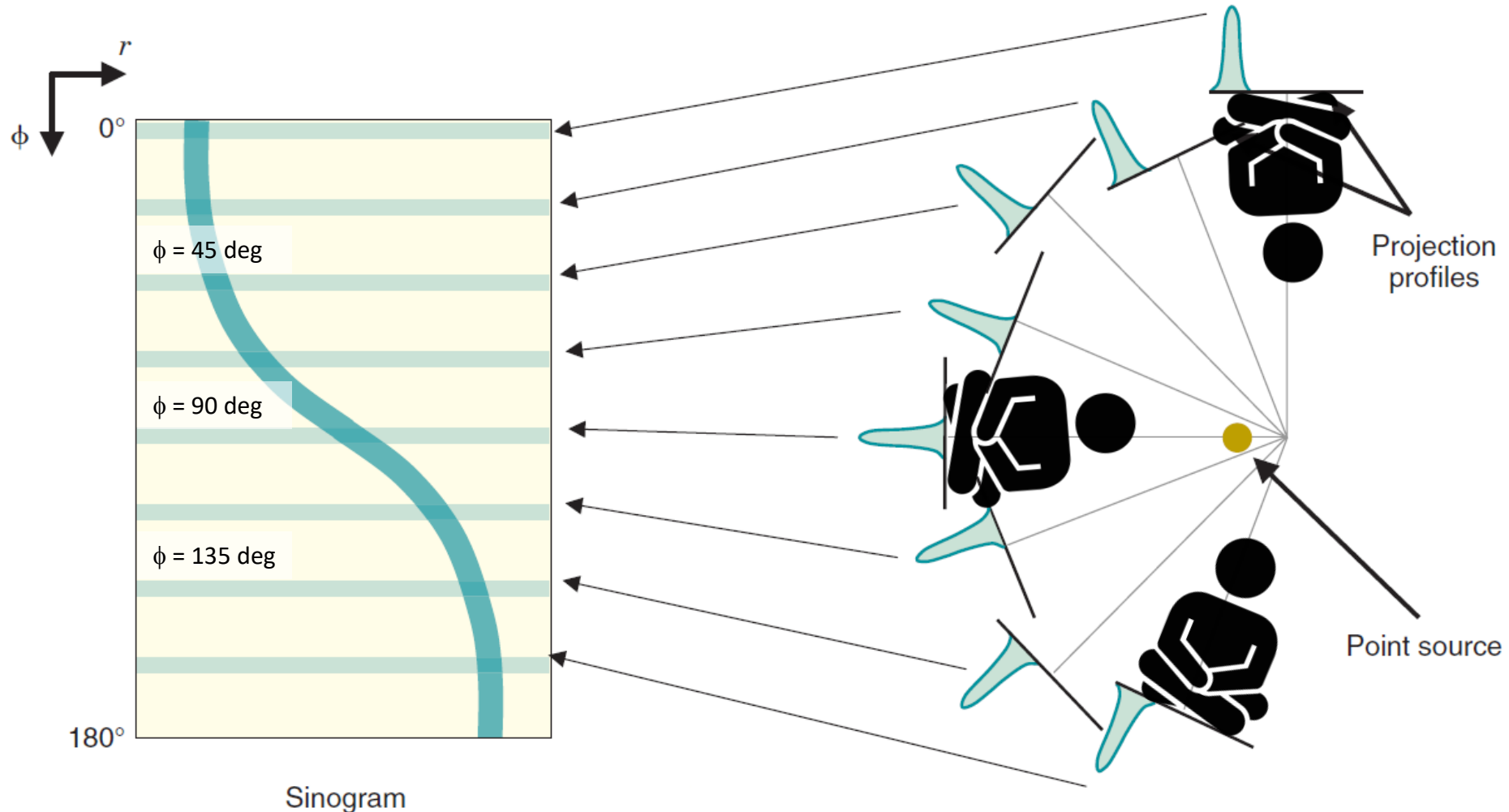
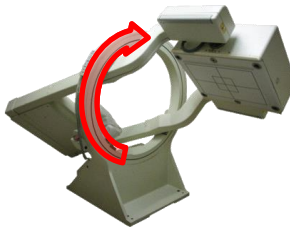
corresponds to each projection angle

corresponds to the position on the camera

1 slice = N projections \leftrightarrow N angular samples \leftrightarrow N sinogram lines

1D Projection data are arranged in a 2D (r, ϕ) **sinogram representation**

Each row (ϕ) in the sinogram displays the intensity profile measured in the corresponding projection



Each **row** (ϕ)

Each **column** (r)

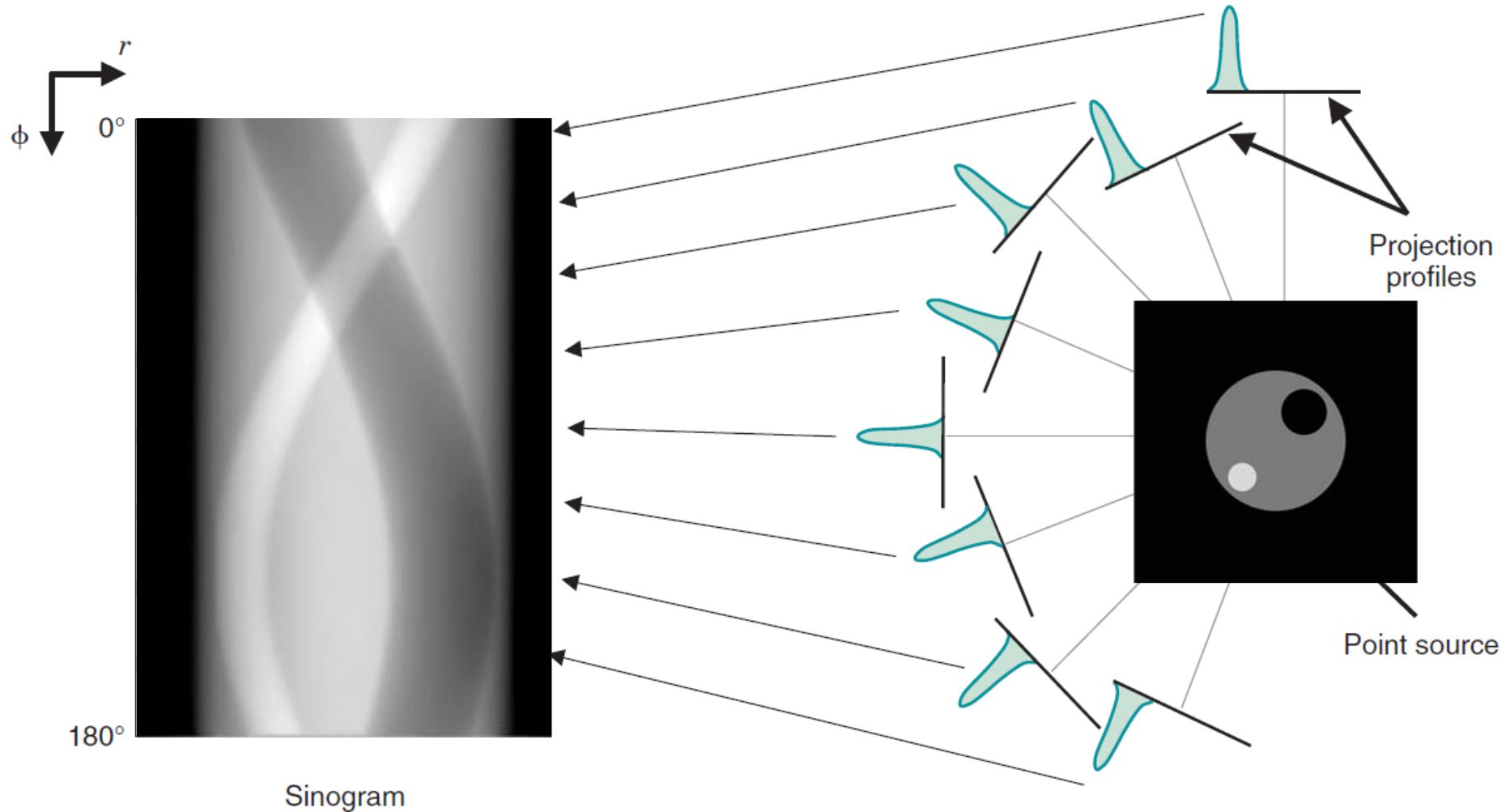
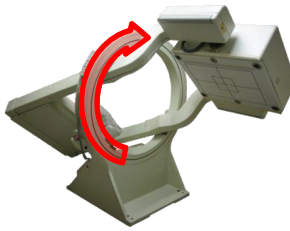
corresponds to each projection angle

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1 slice = N projections \leftrightarrow N angular samples \leftrightarrow N sinogram lines

1D Projection data are arranged in a 2D (r, ϕ) **sinogram representation**

Each row (ϕ) in the sinogram displays the intensity profile measured in the corresponding projection



Each **row** (ϕ)

Each **column** (r)

corresponds to each projection angle

corresponds to the position on the camera

each slice has its own sinogram

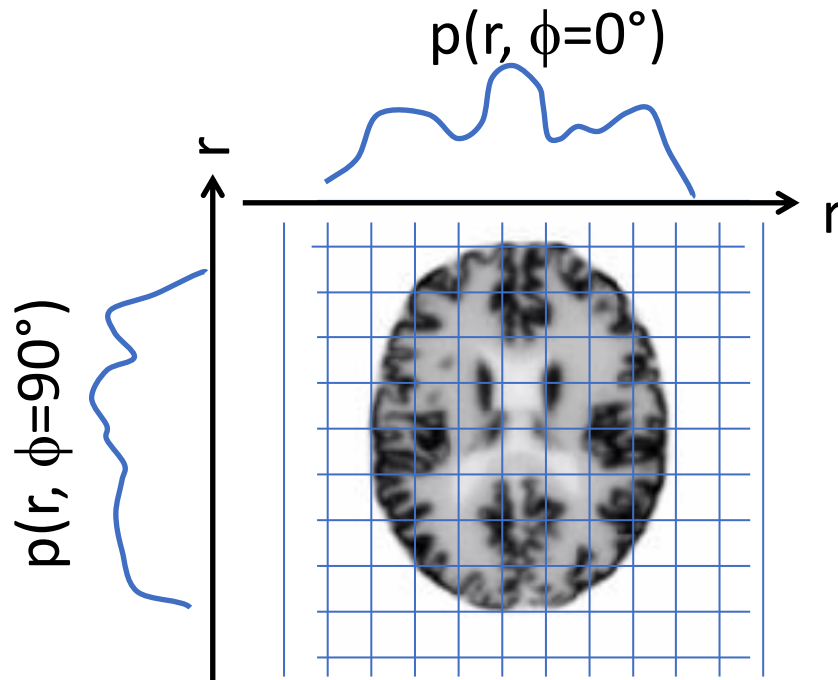
1 slice = N projections \leftrightarrow N angular samples \leftrightarrow N sinogram lines

Reconstruction

What we have:

a discrete number of measured projections $p(r, \phi)$ arranged in a sinogram

What we want: Recover the unknown (true) activity distribution $f(x, y)$!

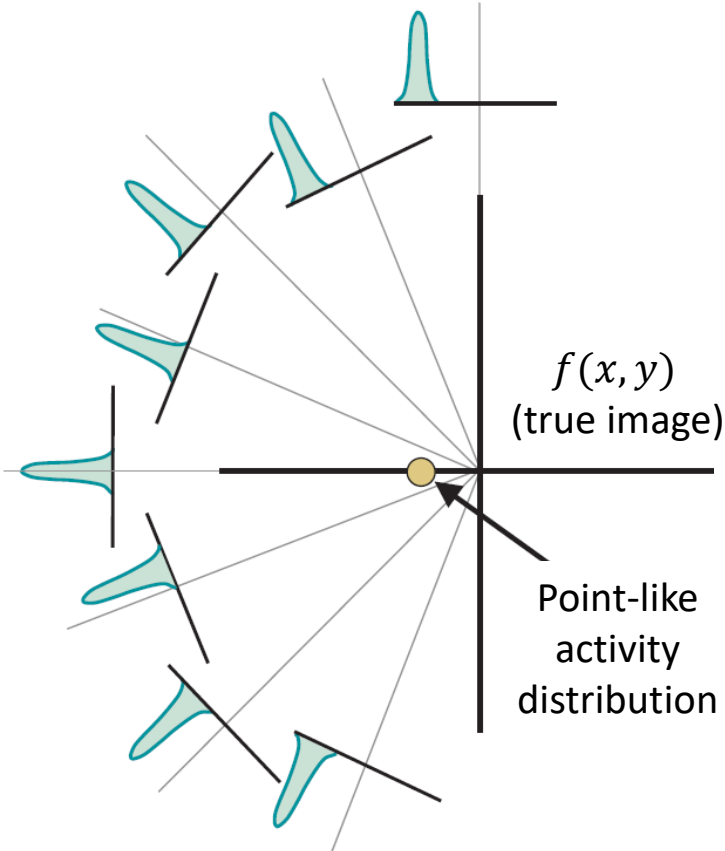


By reconstructing our image, we can access the activity distribution $f'(x, y)$ in a 2D plane of discrete pixels.

Matrix dimension in SPECT is typically 128 x 128

Simple backprojection

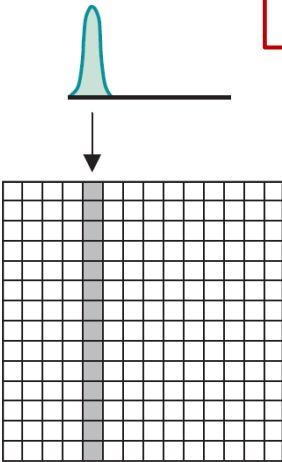
the unknown activity distribution is reconstructed by spreading the measured intensity back across the (pixelized/finite) construction plane.



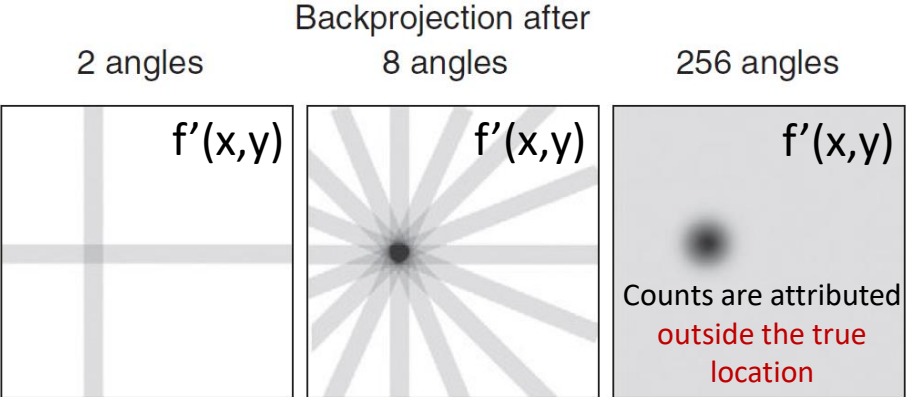
Simple backprojection results in a blurred reconstructed image

1/r blurring

(blurring reduces radially from the centre of the source)



Backprojection of intensity profile at $\phi = 0^\circ$ across the reconstruction plane

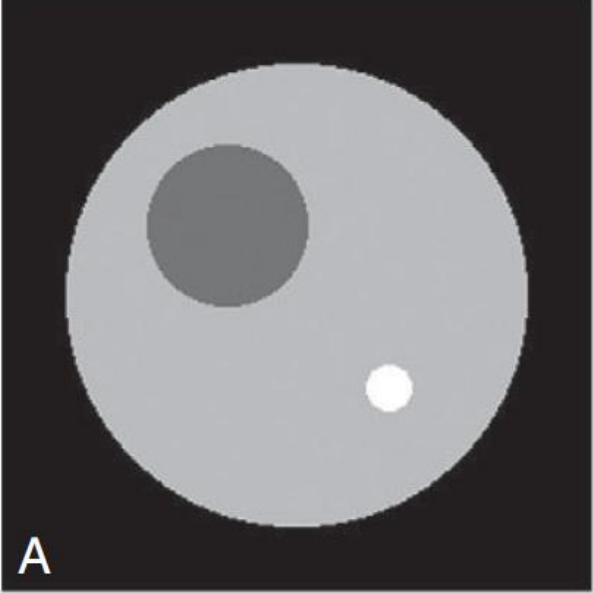


$$f'(x, y) = f(x, y) * (1/r)$$

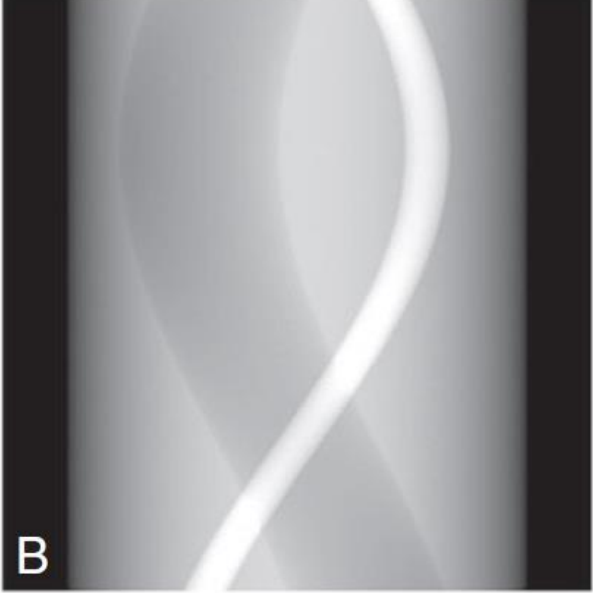
reconstructed source distribution ← original (true) source distribution

Simple backprojection

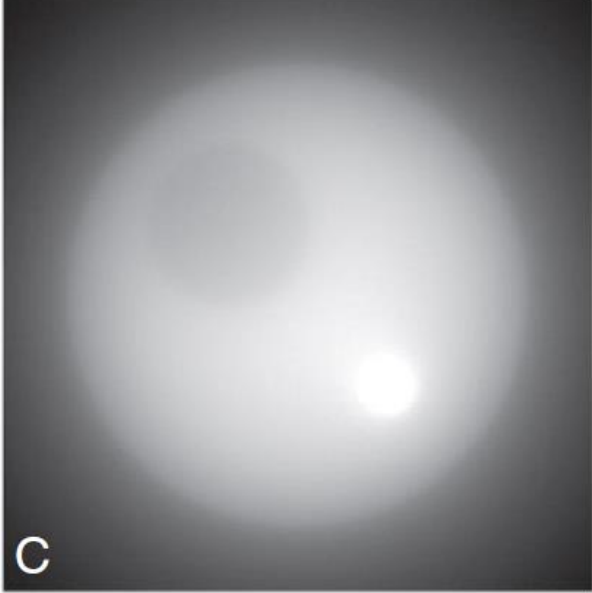
the unknown activity distribution is reconstructed by spreading the measured intensity back across the (pixelized/finite) construction plane.



A Phantom to test reconstruction algorithms (simulated).



B Simulated sinogram (256 projection angles).

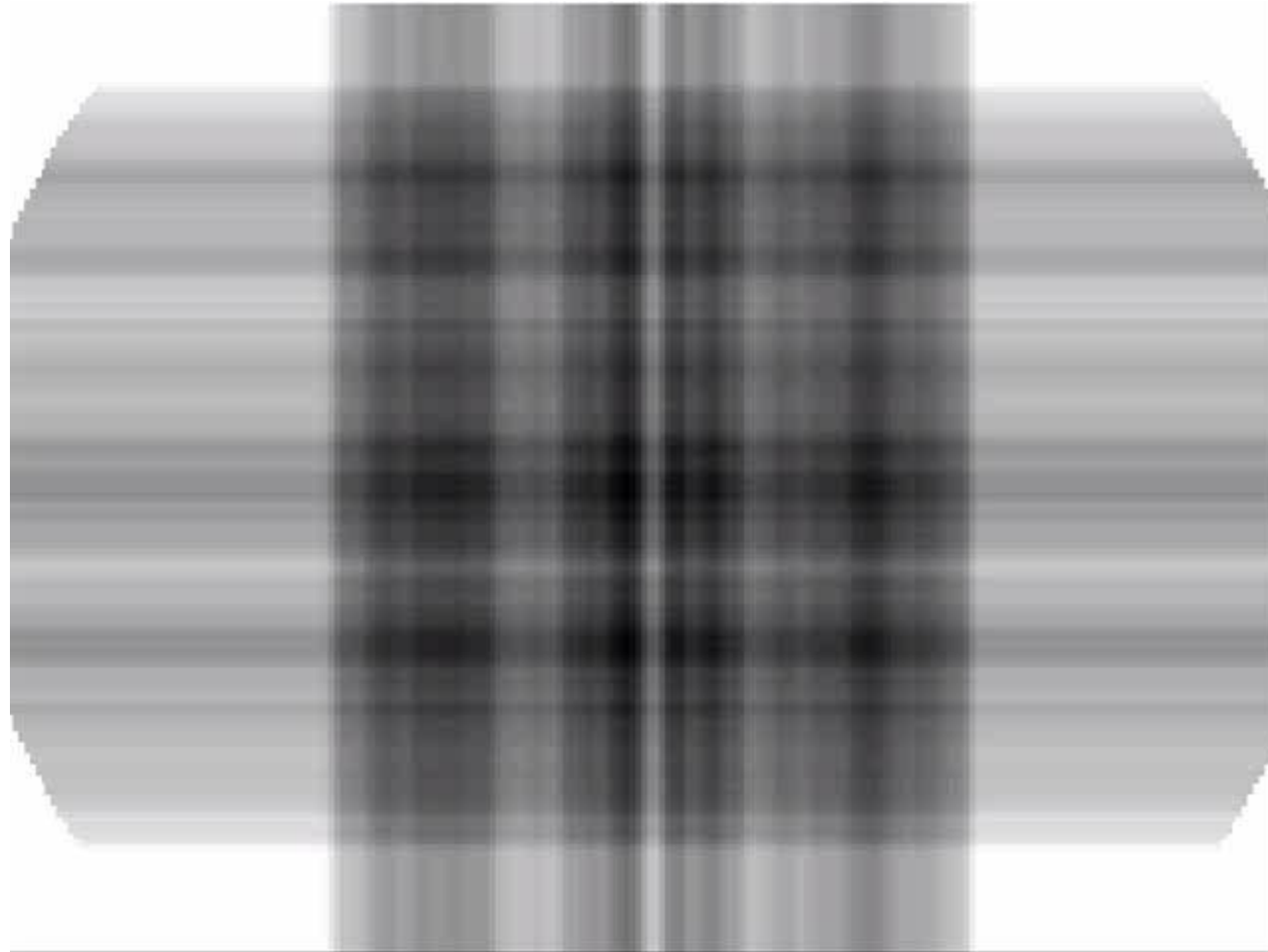


C Simple backprojection reconstruction.

1/r blurring { degradation of sharper details (edges).
degradation of contrast.

Simple backprojection

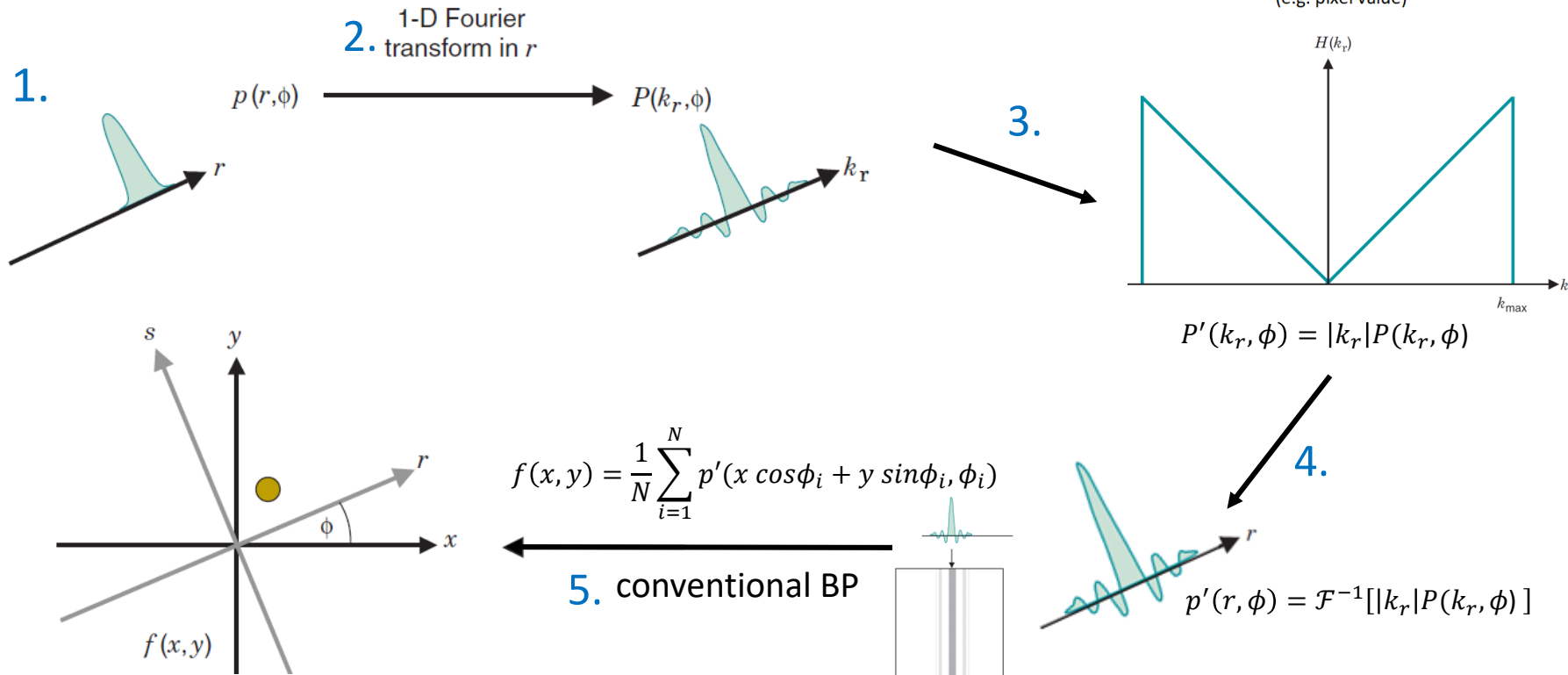
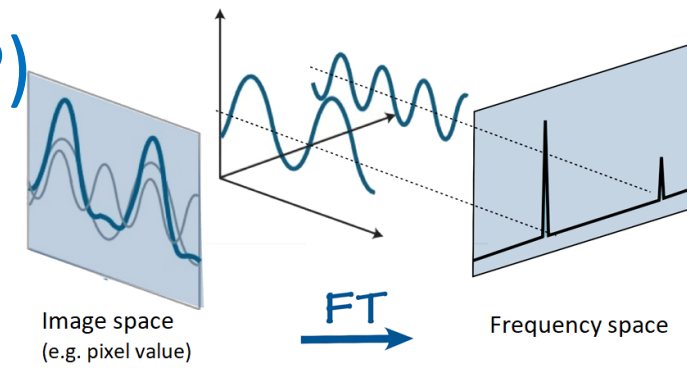
Simple backprojection of an activity distribution in brain



Simple backprojection results in a blurred reconstructed image

Filtered backprojection (FBP)

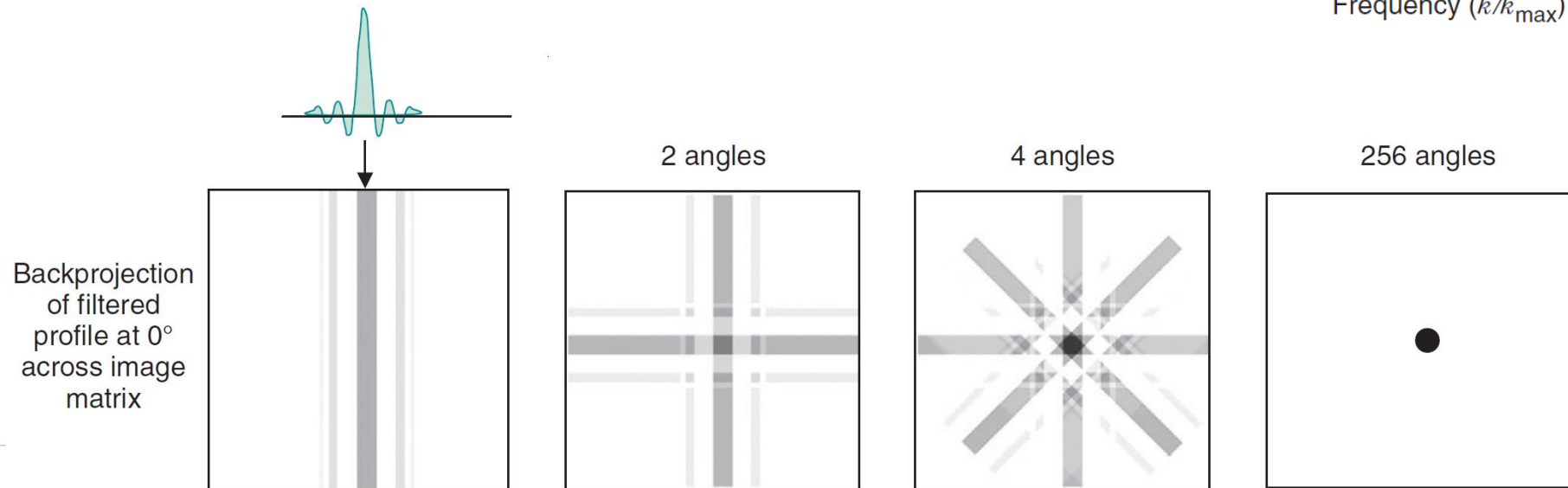
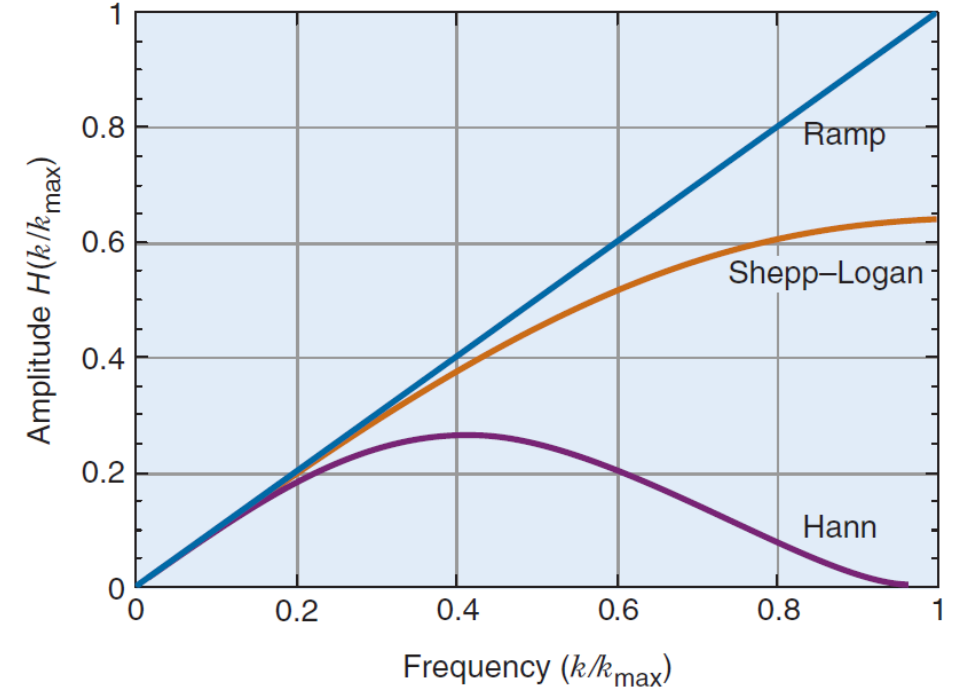
- Image (real) and frequency space are related by the Fourier transform (FT).
- FT provides another method to represent spatially varying data:
 - 1D image profile can be represented as Σ of cos and sin with \neq spatial frequencies k .



1. Acquire N projection profiles $p(r, \phi)$ (sinogram).
2. Compute the FT of each profile (sinogram row).
3. **Apply a filter to each profile in frequency k -space $\rightarrow |k_r|$.**
4. Compute the inverse FT of each **filtered profile** to obtain a **modified projection profile**.
5. Perform **conventional backprojection** with the new (filtered) profiles to compute the image.

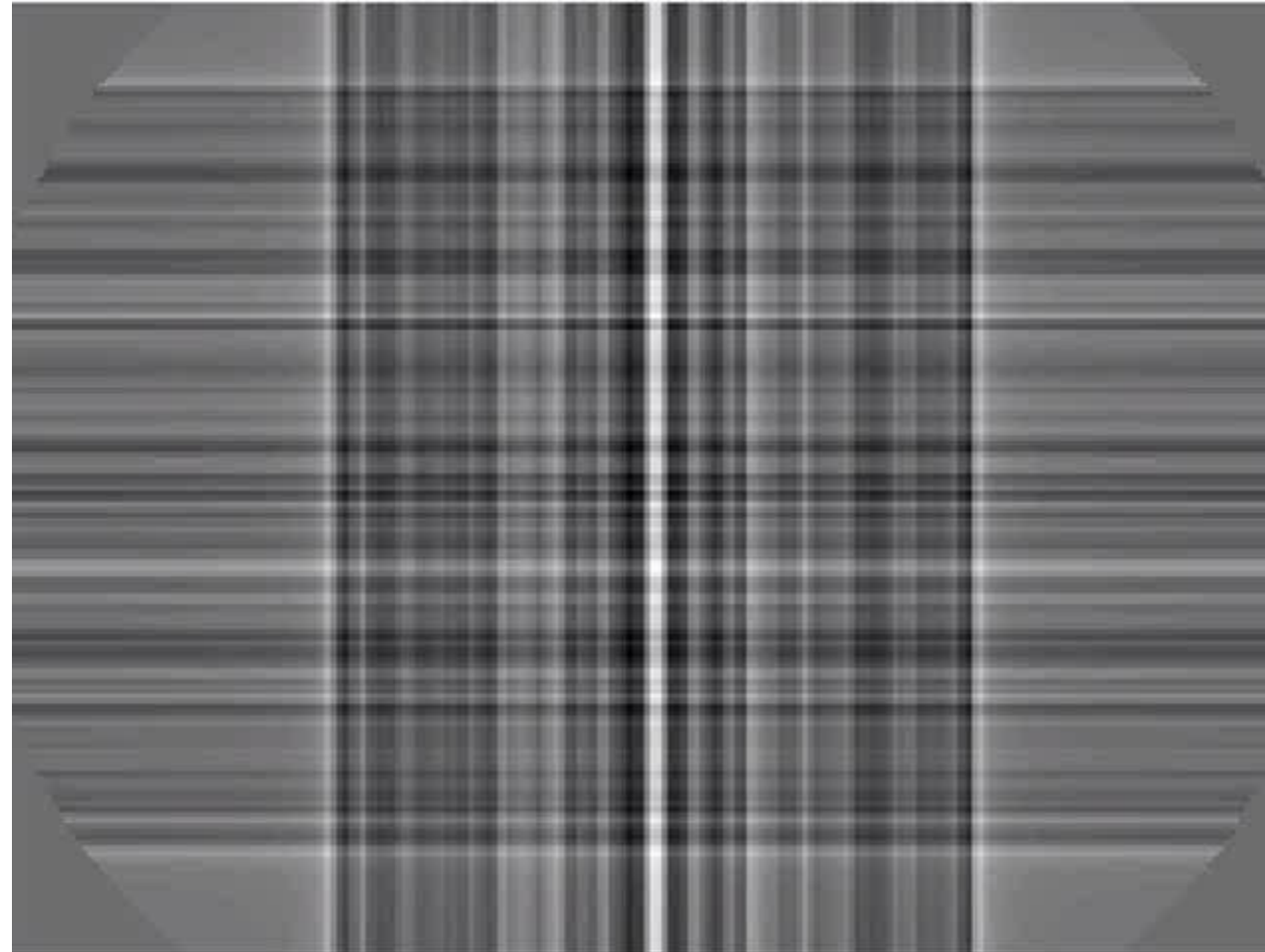
Filtered backprojection (FBP)

- Filtering in k -space precedes backprojection into object (image) space.
- Ramp filter
 - ✓ Low frequency damping
 - ✓ Removes $1/r$ blurring + sharpens detail.
 - High frequency selected
 - ✗ High frequency noise amplified.
 - ✗ Degradation of signal to noise ratio (SNR).
- FBP images appear sharper but noisier than simple BP.
- Other filters can be used to mitigate this effect (“rounded shape” filters).



Filtered backprojection (FBP)

Filtered Backprojection (FBP) of an activity distribution in brain



If noiseless projections and virtually infinite angles views are available



Quantitative true activity distribution reconstruction

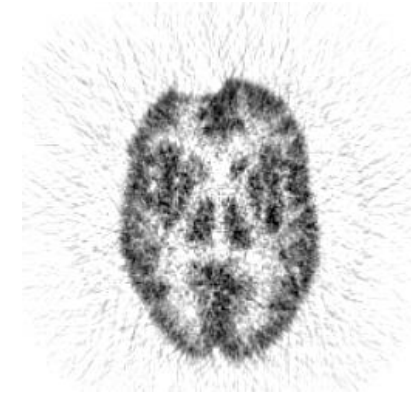
Filtered backprojection (FBP)

Advantages :

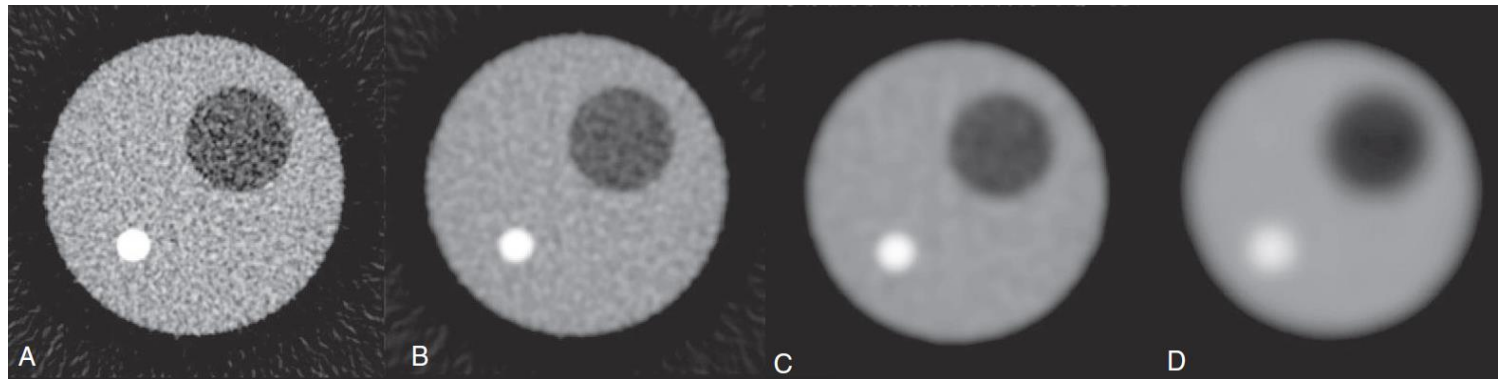
- Accurate if noise-free measurements and completely sampled data.
- Fast and easy to implement.

Limitations :

- Major artifacts if data is measured incompletely.
- Still noisy → can only be partially mitigated by filters.
- No natural way to implement corrections for:
 - Spatial resolution
 - Scattering
 - Attenuation



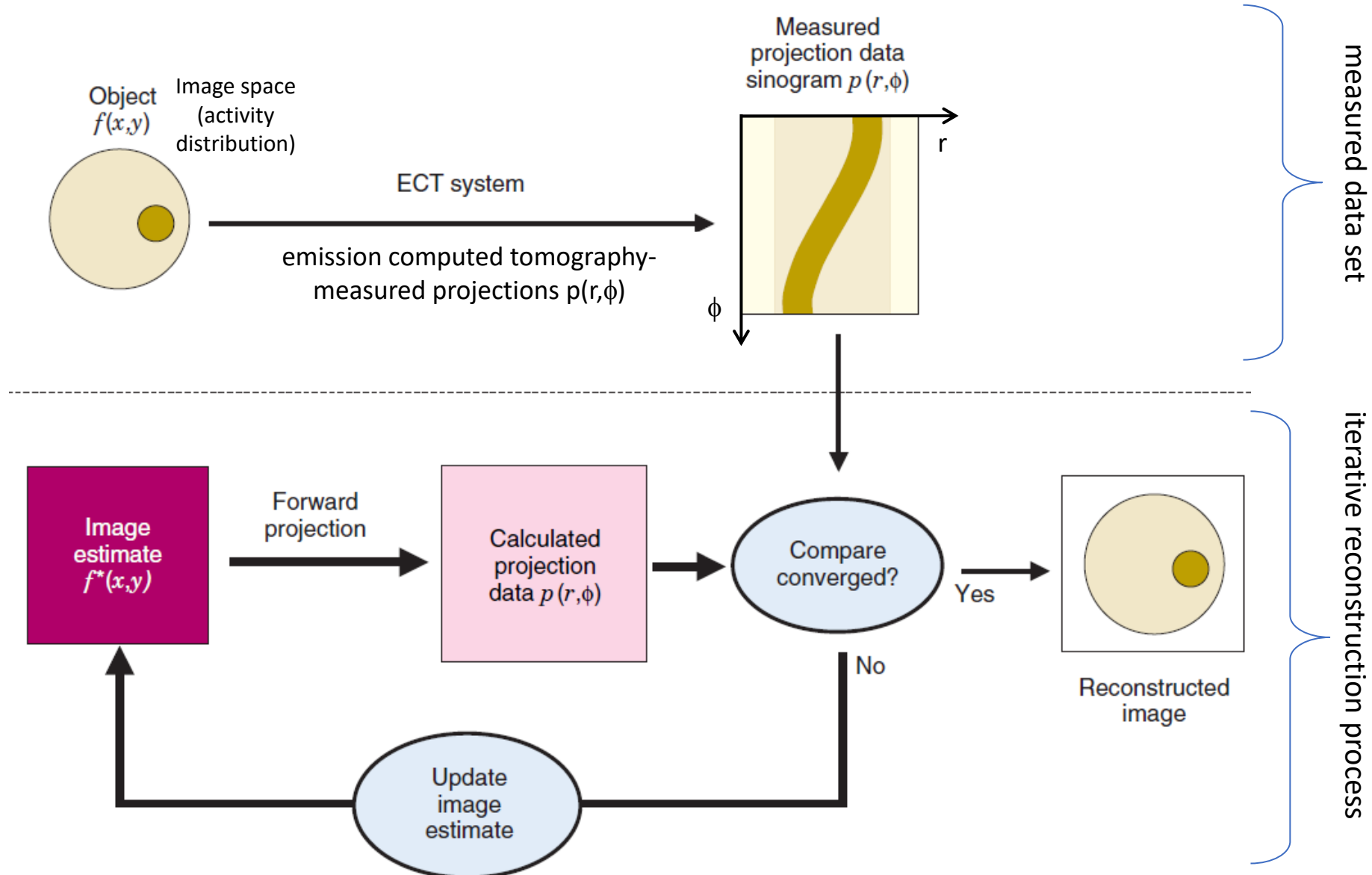
iterative reconstruction techniques !



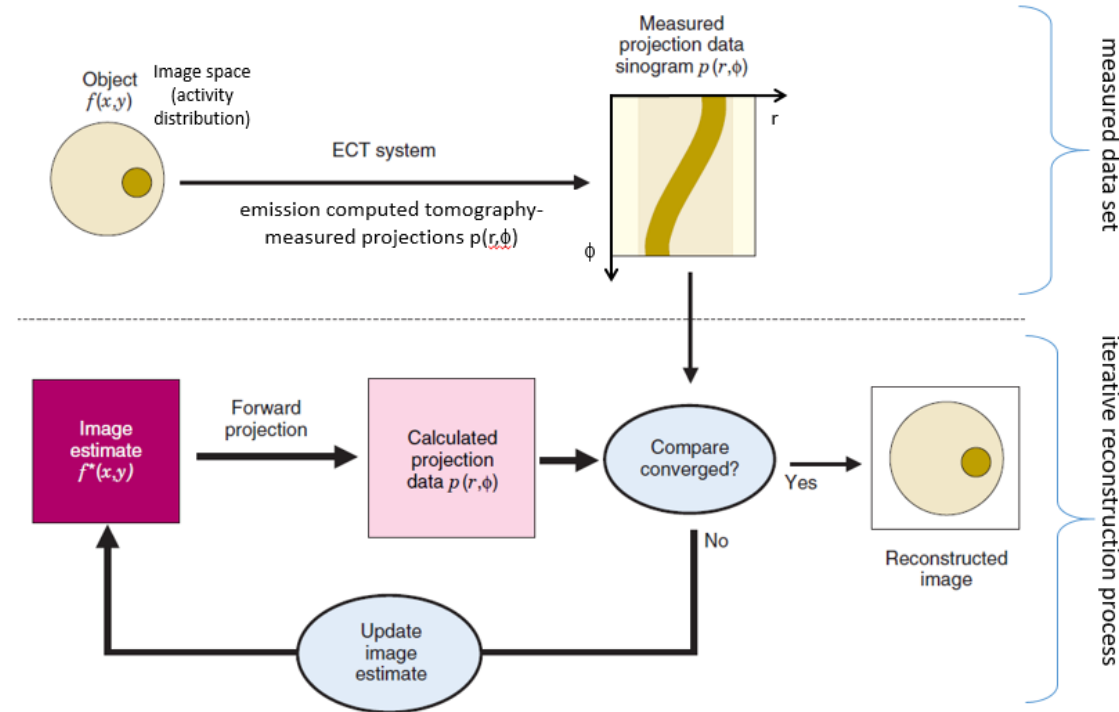
Example: FBP with Shepp-Logan filter and different cutoffs → tradeoff between SNR and image sharpness/detail.

Iterative reconstruction

more computationally intensive than FBP, but **standard nowadays**.



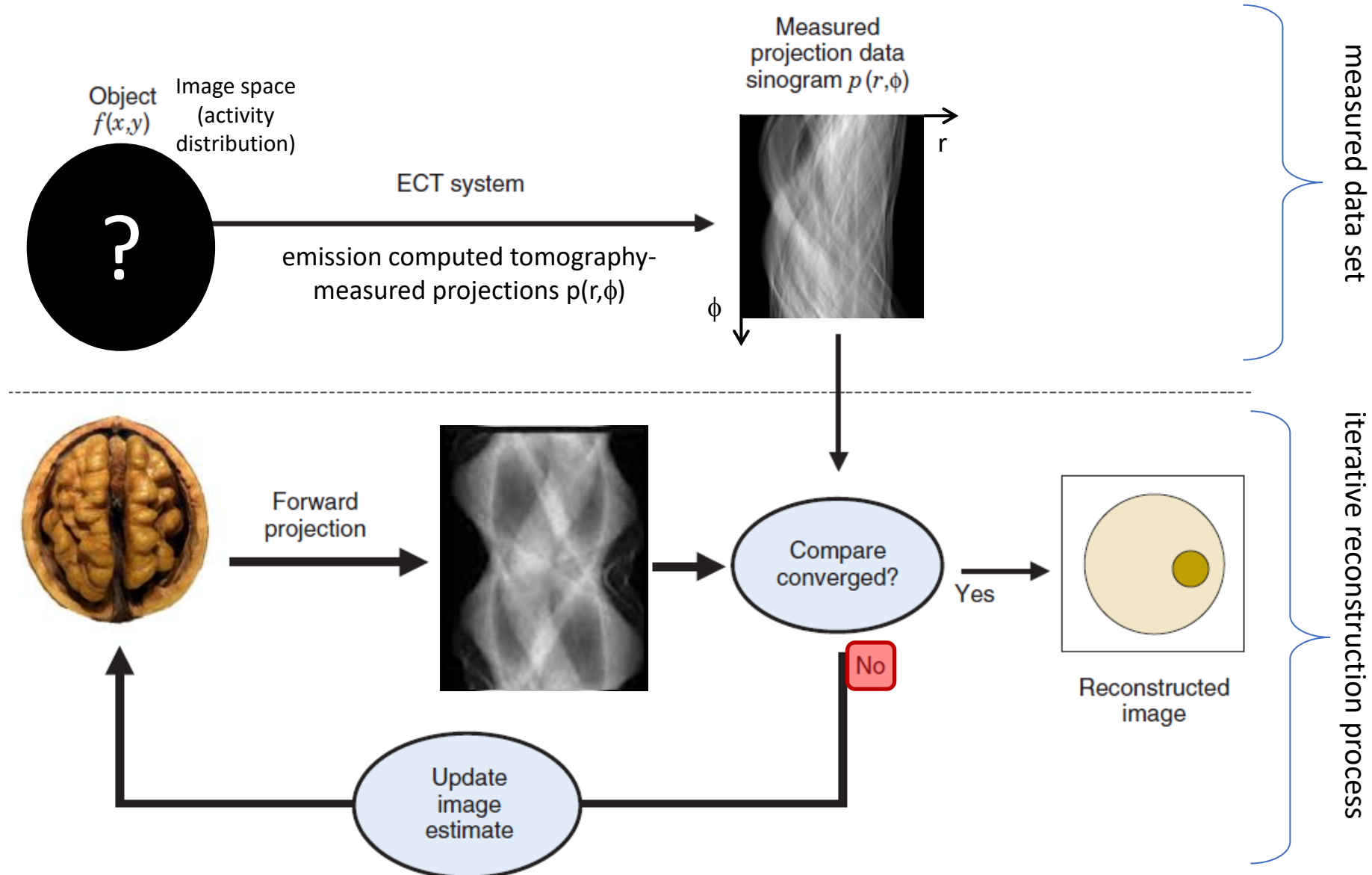
Iterative reconstruction



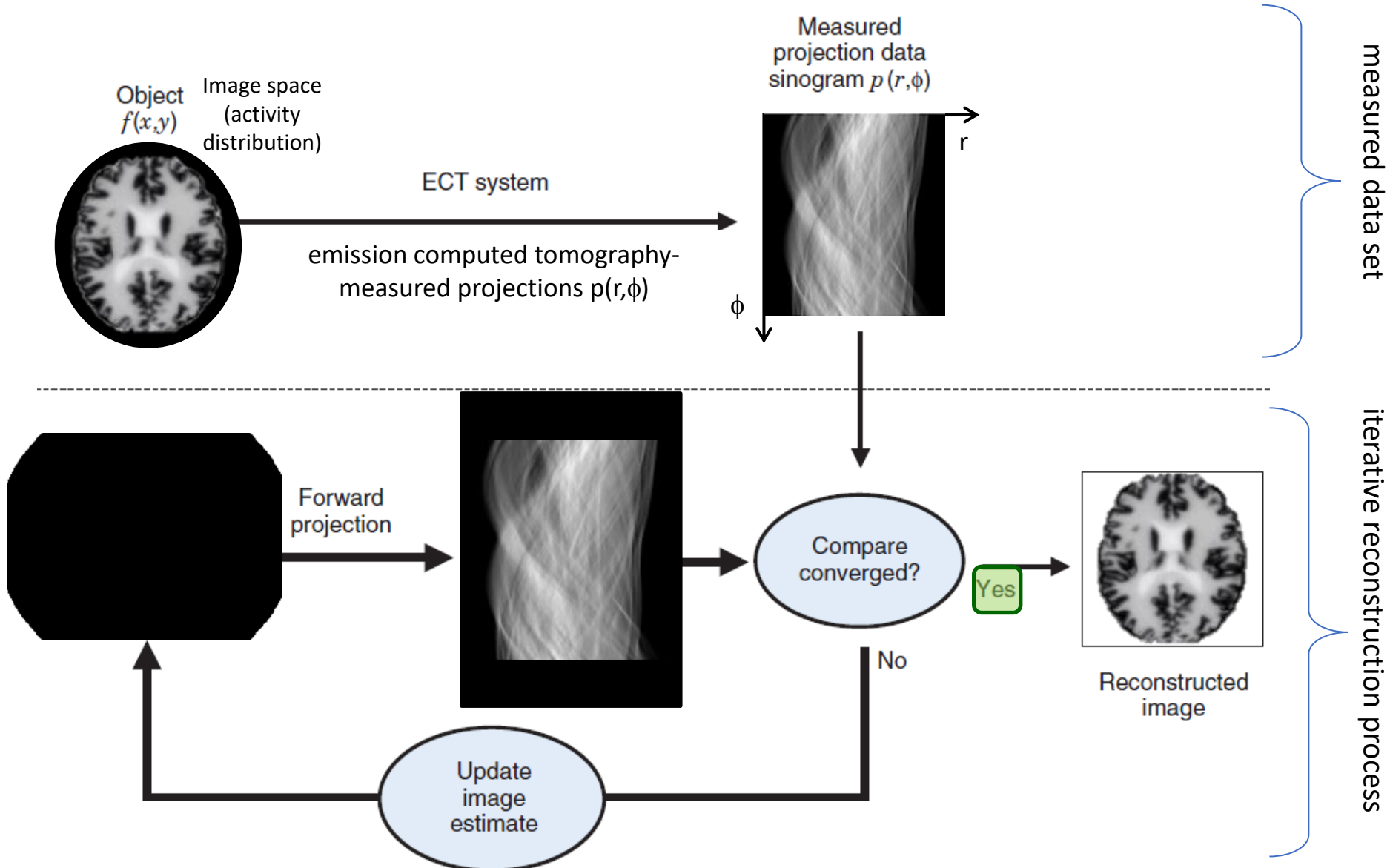
True activity distribution $f(x,y)$ is estimated by successive approximations $f^(x,y)$:*

1. Start with a simple initial image estimation $f^*(x,y)$ (e.g. uniform activity distribution in the FOV or FBP).
2. Compute (forward) projections to calculate estimated sinogram.
3. Compare measured and computed sinogram.
4. If no convergence is reached, compute correction factors.
5. Refresh the image estimate and continue iterating until convergence.

Iterative reconstruction



Iterative reconstruction

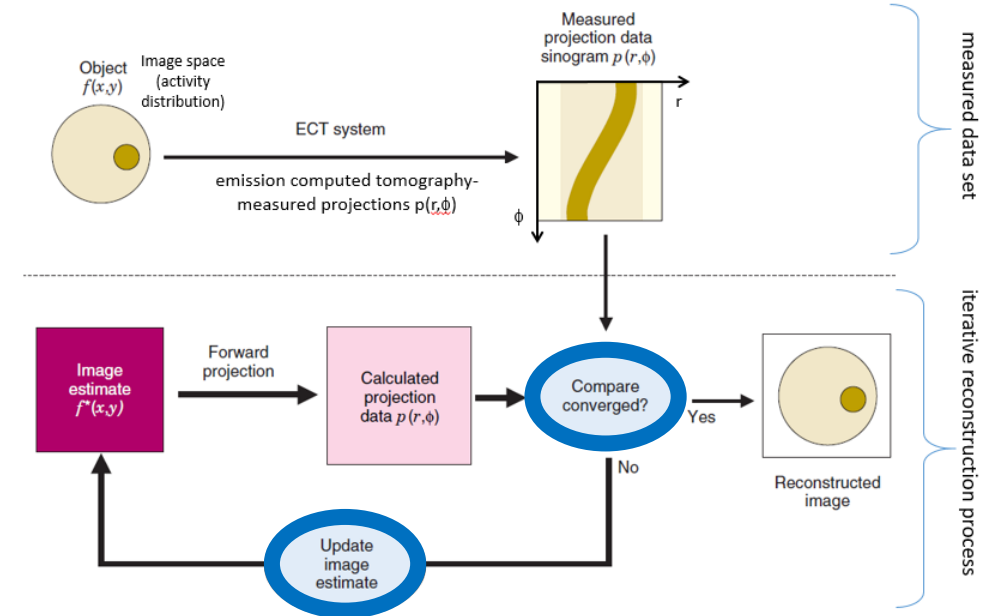


Iterative reconstruction

Iterative reconstruction is based on:

1. Comparing estimated vs. measured data
2. Updating the image estimate

→ **Maximum-likelihood expectation maximization algorithm (MLEM)**
Compute the maximum-likelihood source distribution that would result in the measured projection data (sinograms).



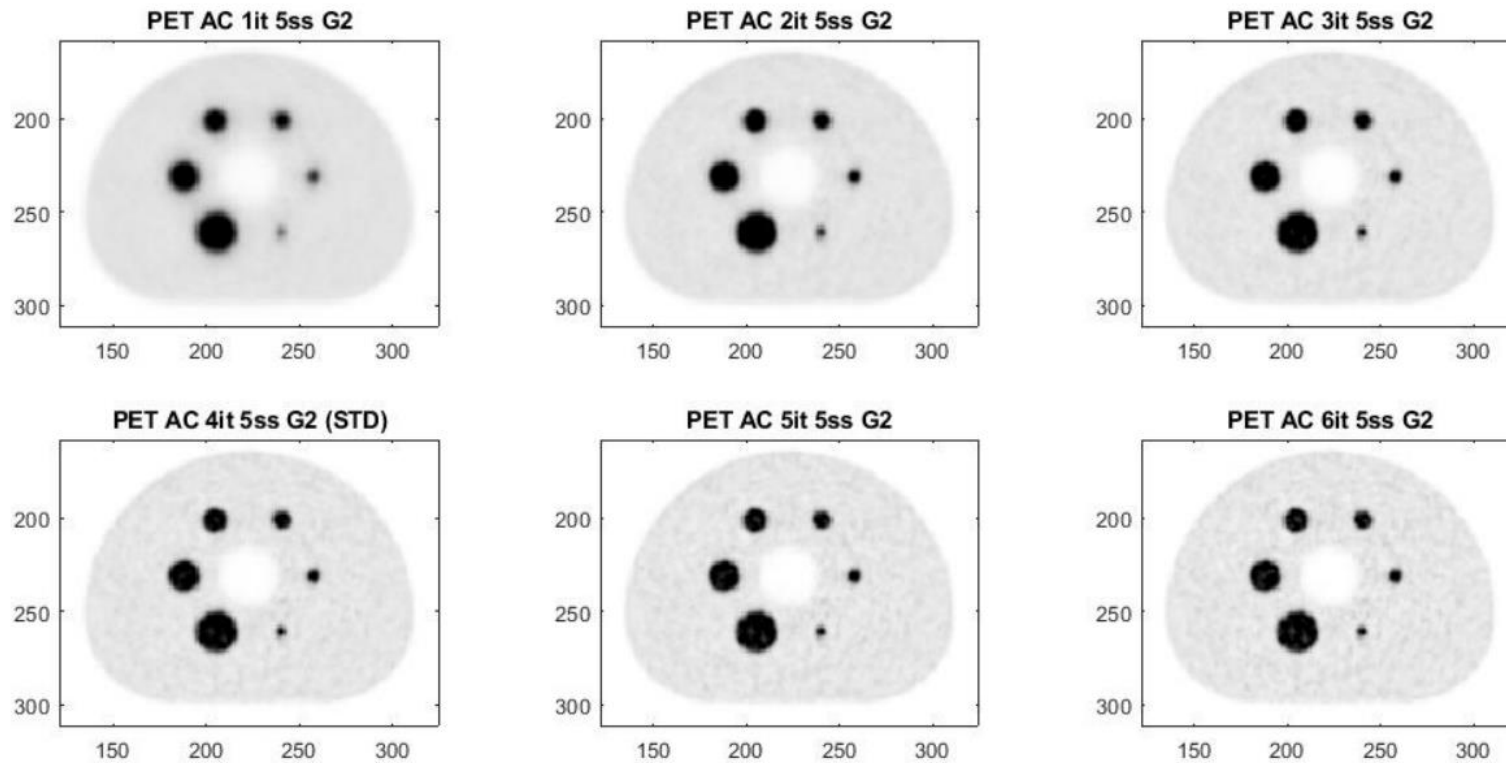
Iterative reconstruction is more computationally intensive than FBP:

- Needs several iterations to converge.
- Reconstruction algorithms allow to include specific characteristics of the imaging modality:
 - attenuation,
 - scatter,
 - finite resolution, system geometry.
- Speed-up solution → ordered subsets (OSEM) : only a small number of projections (angles) are used in the first iteration steps (time per iteration \propto # projection profiles to compute).

Iterative reconstruction = nowadays standard

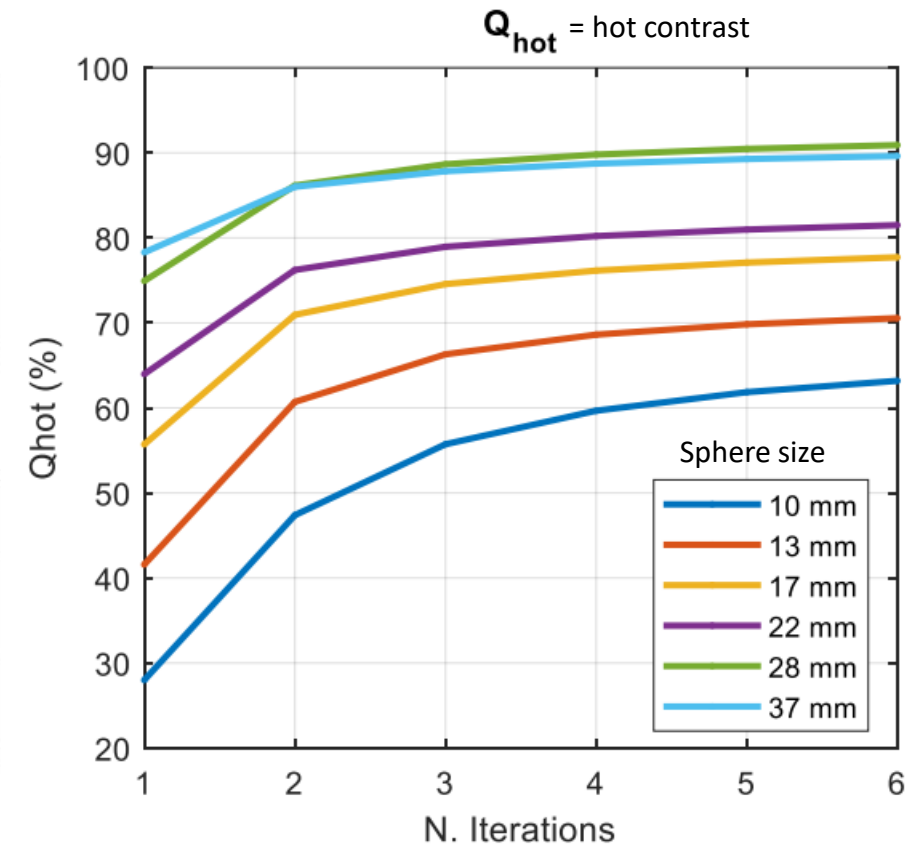
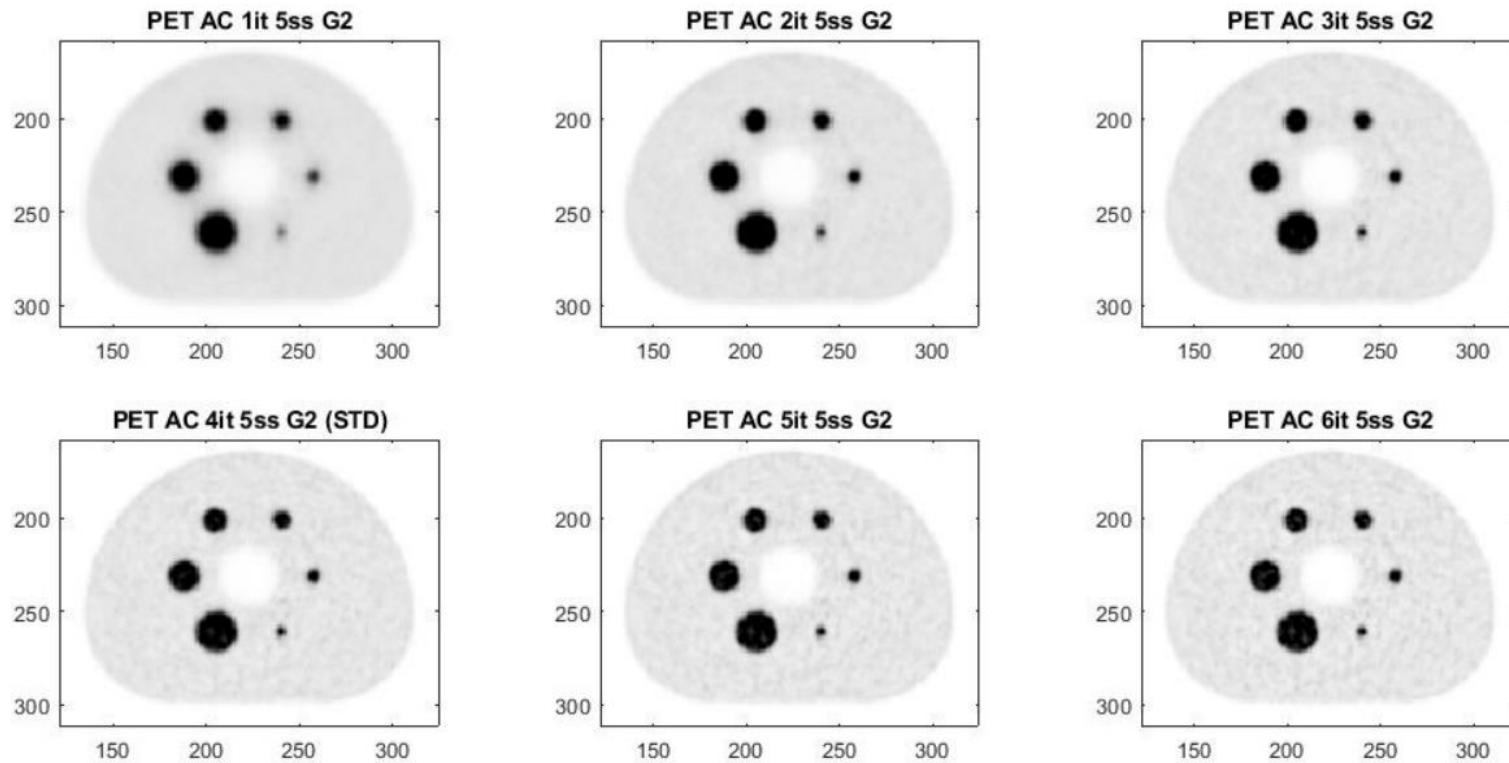
Small structures need more iterative step to reach convergence!
(but noise amplification with # iterations)

Test object :
phantom with spherical inserts



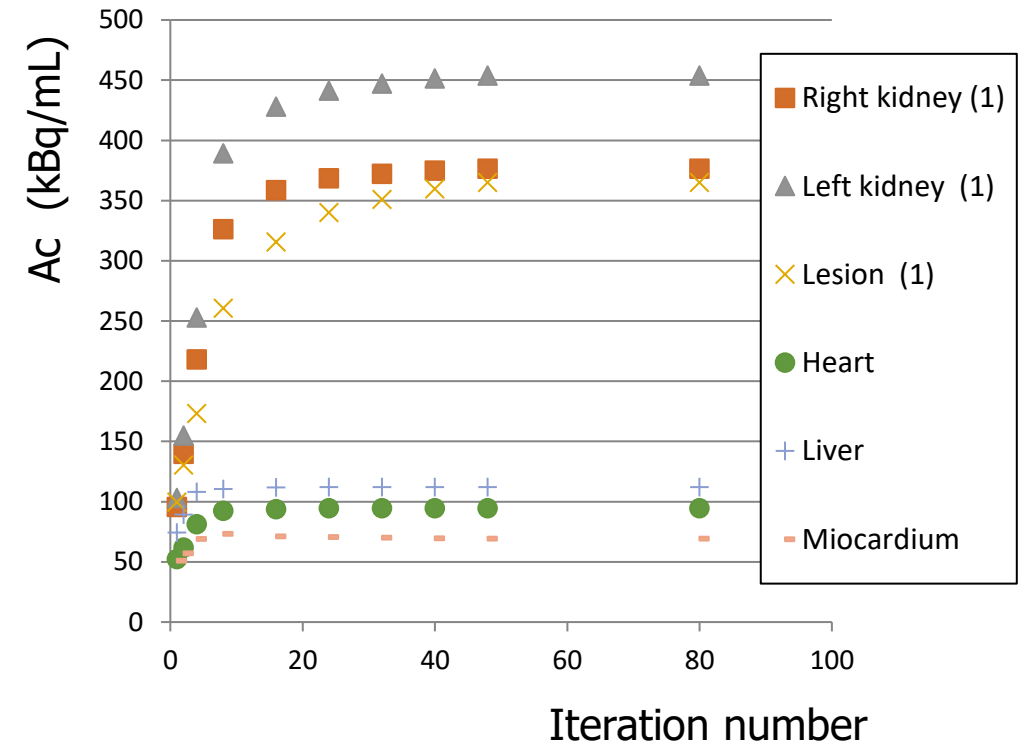
Iterative reconstruction = nowadays standard

Small structures need more iterative step to reach convergence!
(but noise amplification with # iterations)



Iterative reconstruction = nowadays standard

Small structures need more iterative step to reach convergence!
(but noise amplification with # iterations)



Single Photon Emission Computer Tomography: SPECT



Single head



Double head

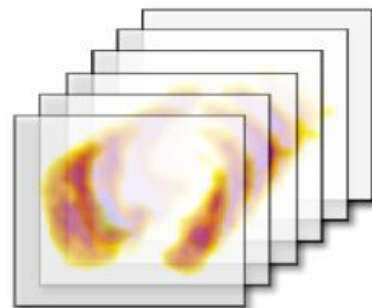


Triple head

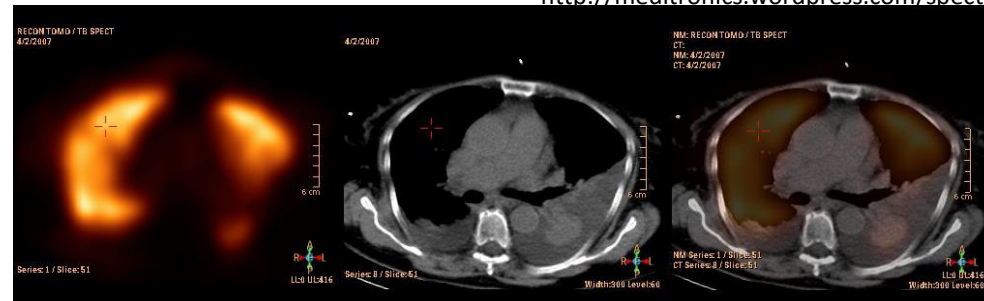
Acquisition: 2D array of 1D-projections
→ 2D planar image

Tomographic
reconstruction

2D axial slices
stacked in a 3D array
(3D volume image)

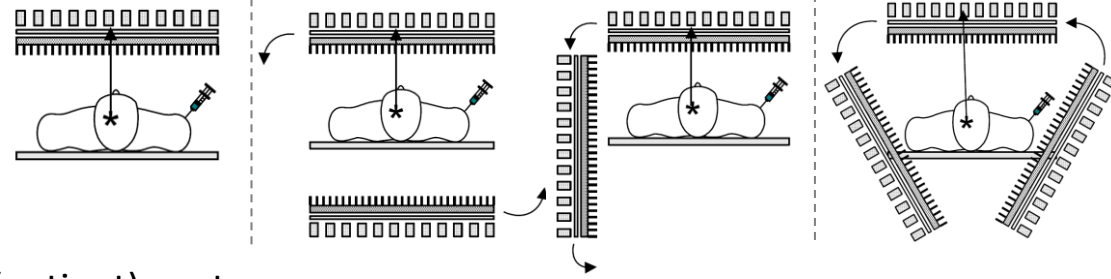


<http://meditronics.wordpress.com/spect/>



Multiple detection heads

- increased detection sensitivity
- reduced acquisition time

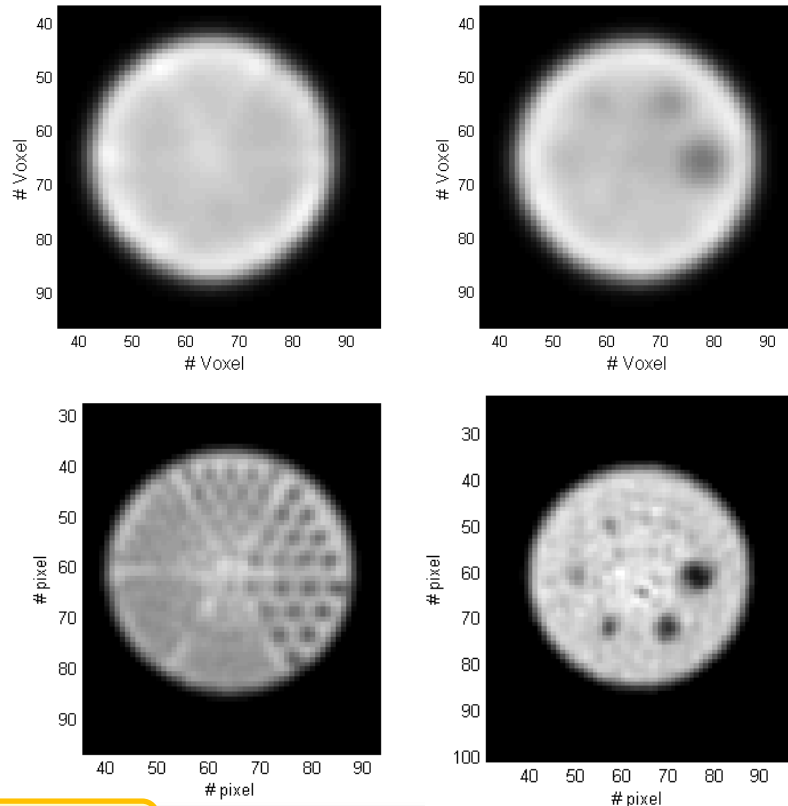


Acquisition orbit

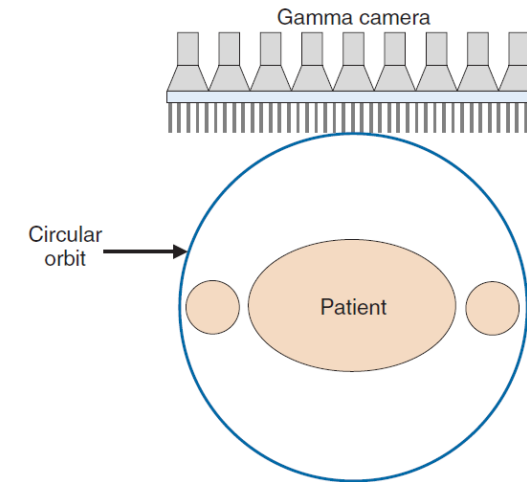
- as close as possible to the object (patient) contour
- minimize resolution degradation with distance



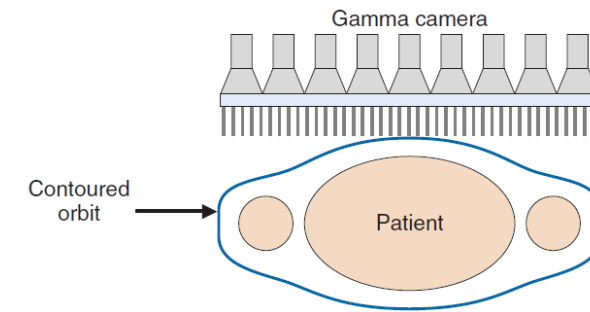
Automatic patient detection systems



fixed circular orbit
 $\varnothing \sim 60$ cm



automatic
adaptive orbit
 $\varnothing \sim 34$ cm



Special designs: neuro, cardiac and preclinical imaging

SPECT

Source: Quality Assurance in Gamma Camera & SPECT Systems, James R. Halama

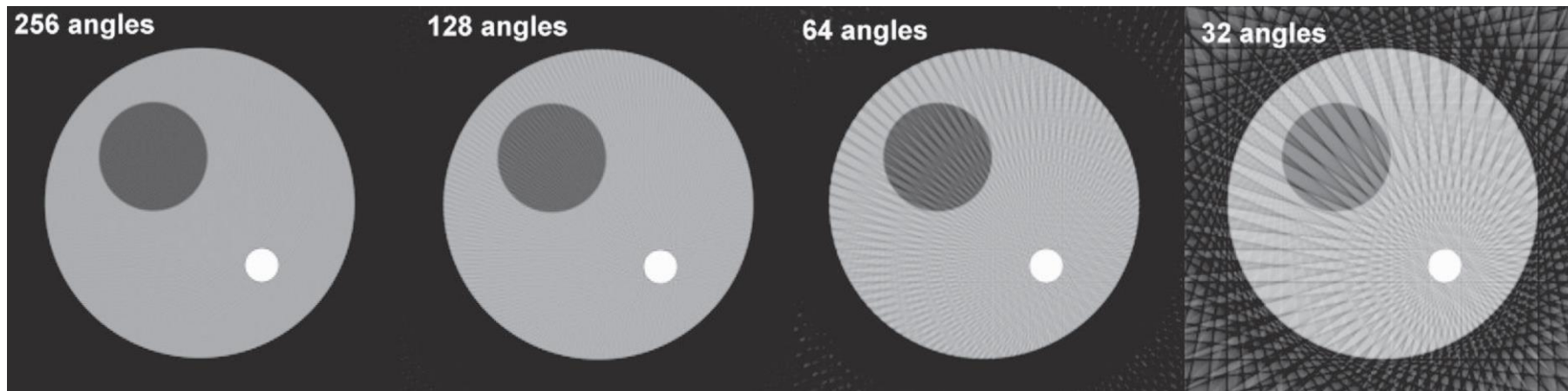
83

Source: Physics in Nuclear Medicine, 4th edition, ISBN: 978-1416051985

Artefacts

how projections should be acquired ?

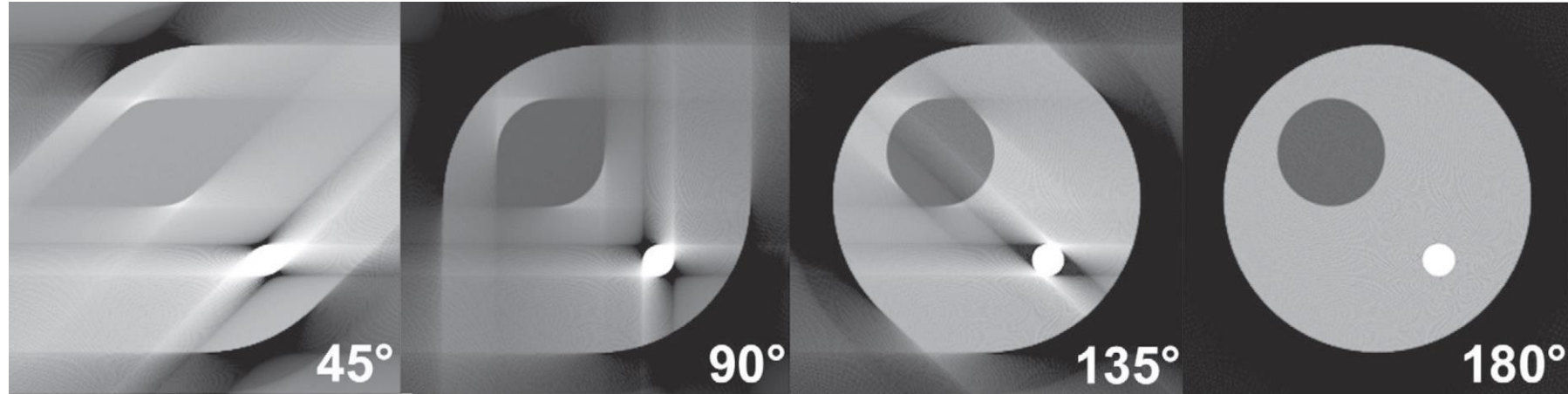
Angular undersampling (number of sampled angles) → streak-like artifacts.



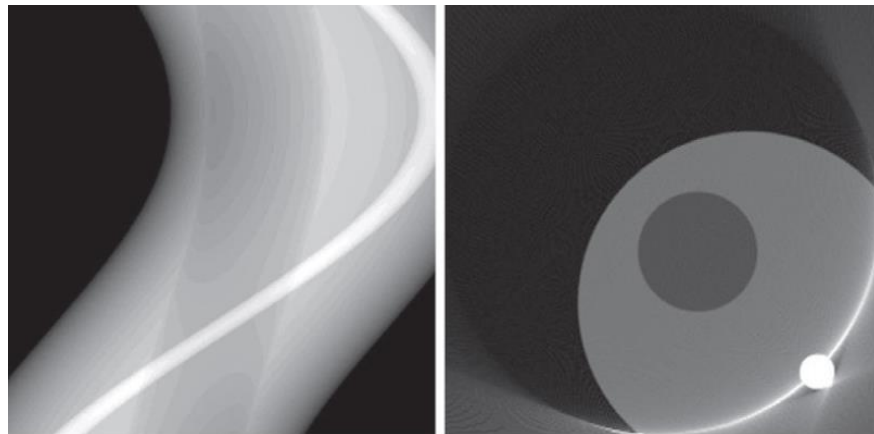
Artefacts

how projections should be acquired ?

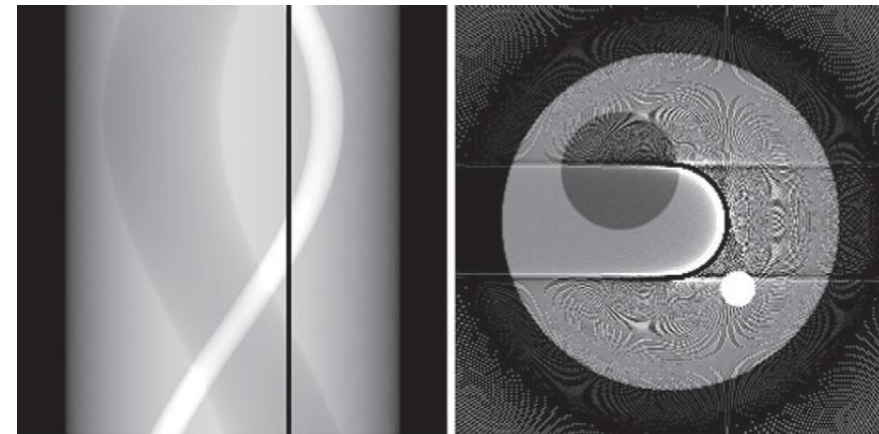
Projections acquired over less than 180° \rightarrow artifacts.



Projections not covering the whole object \rightarrow artifacts

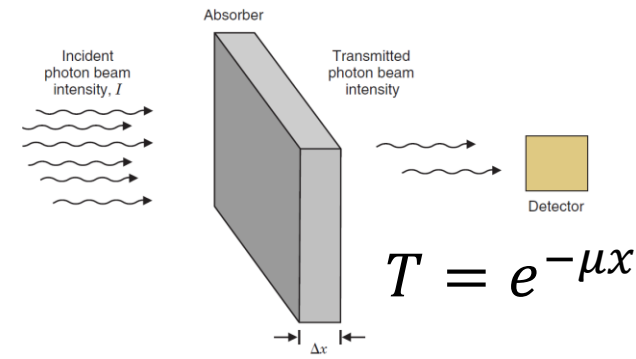


Missing projection (part of detector kaputt) \rightarrow artifacts



Attenuation

- Attenuation is the inverse of transmission.
- It depends on the distance the photons need to travel through the tissue.
- Attenuation coefficient μ depend on **density**, atomic number **Z** and **energy**.



μ : linear attenuation coefficient (in cm^{-1})

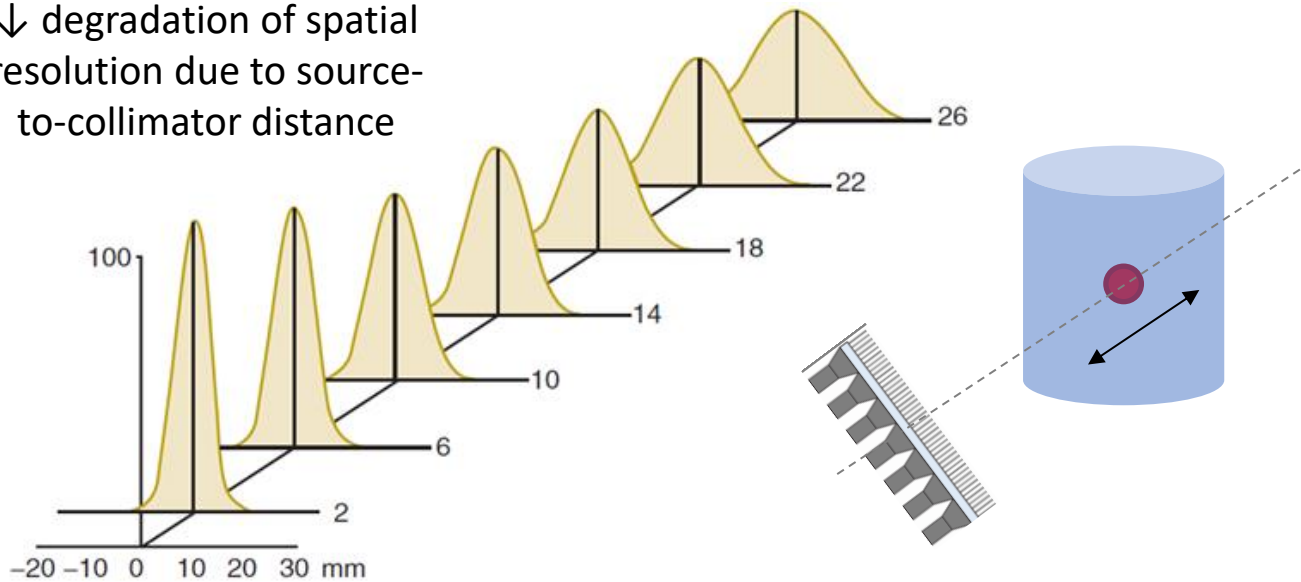
$\mu_{\text{soft tissue}}(140 \text{ keV}) = 0.155 \text{ cm}^{-1}$

$\mu_{\text{bone}}(140 \text{ keV}) = 0.25 \text{ cm}^{-1}$

Effect of attenuation :

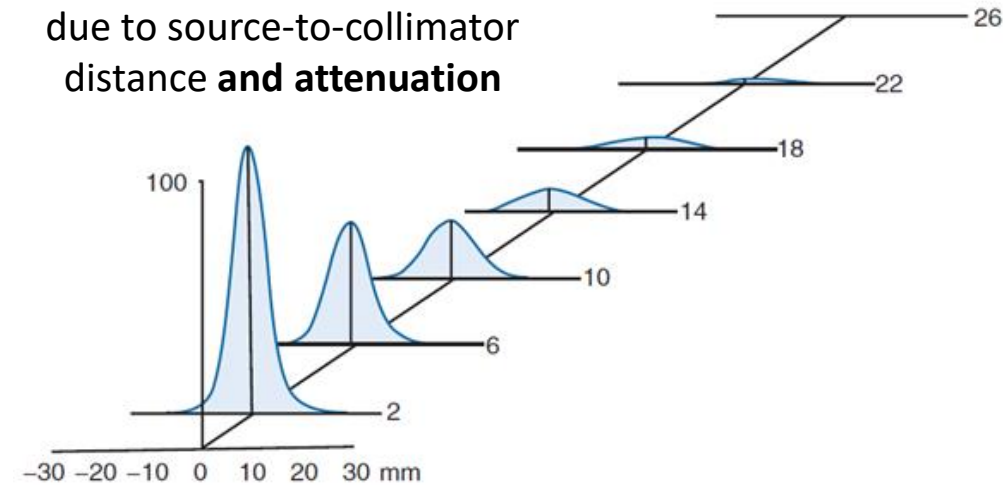
⁹⁹Tc-line source response in air

↓ degradation of spatial resolution due to source-to-collimator distance



⁹⁹Tc-line source response in water

↓ degradation of spatial resolution due to source-to-collimator distance **and attenuation**

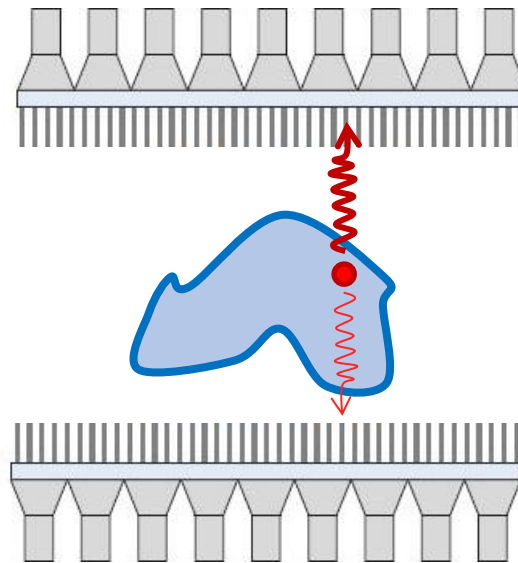
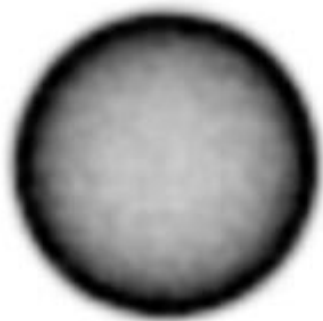


Attenuation correction

Expected

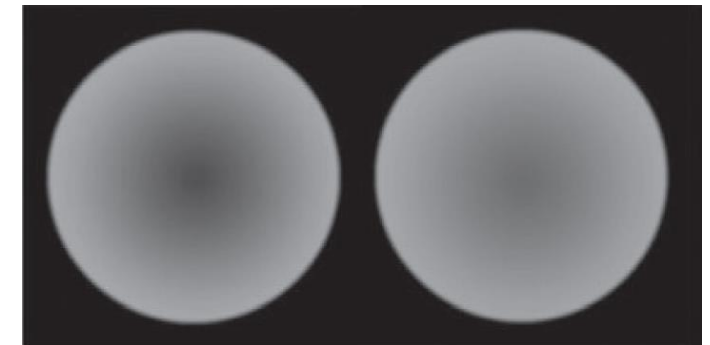
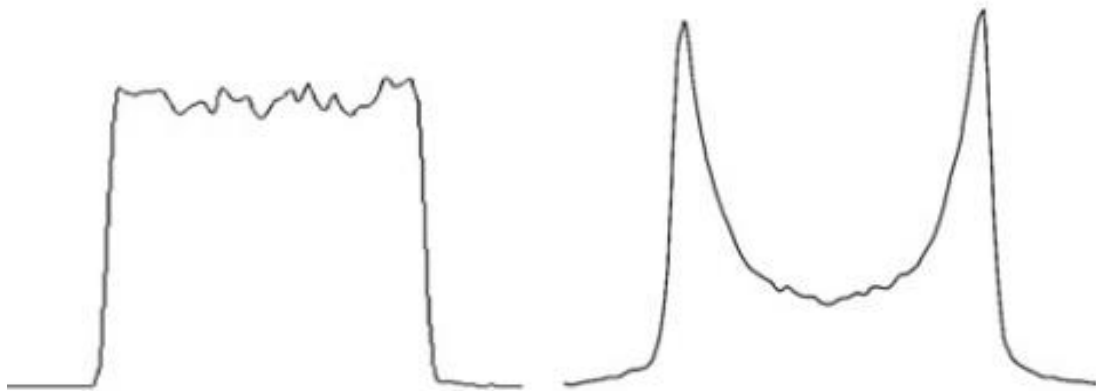


Measured (NAC)



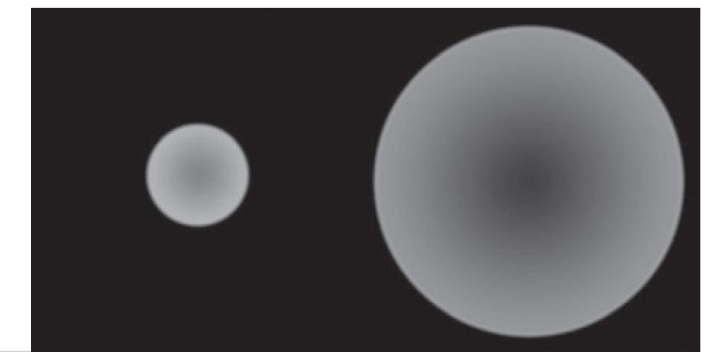
The attenuation correction is influenced by:

- the energy of the photons
- the size/density of the structure to be crossed



^{99m}Tc

^{131}I



10 cm

30 cm

Attenuation correction :

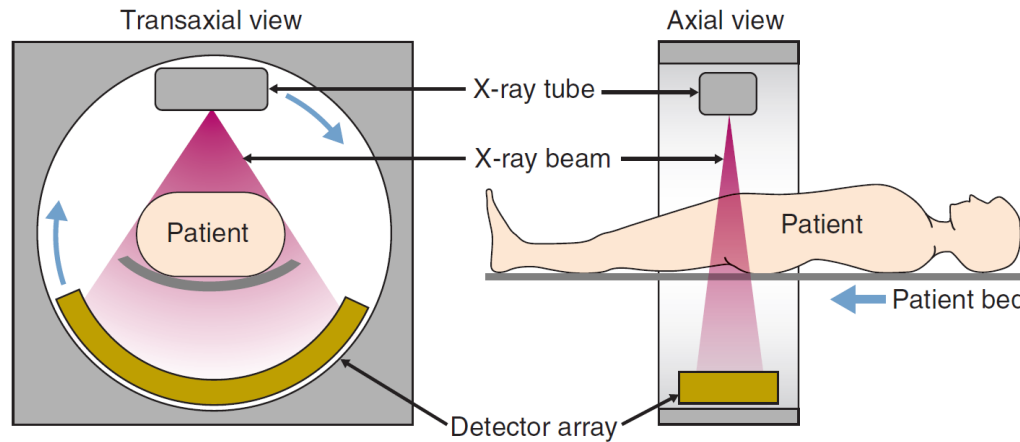
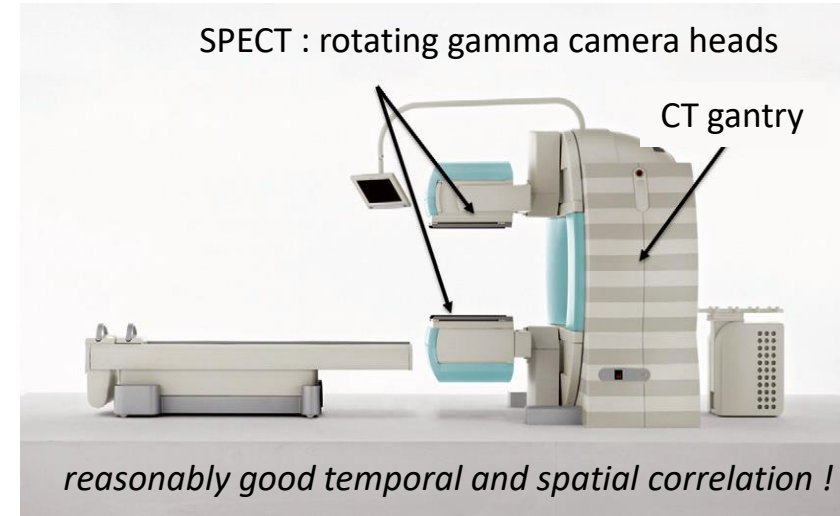
- Corrects (brings back) the pixel values according to the attenuation that the photons experience on their way.

Hybrid systems and attenuation correction (AC)

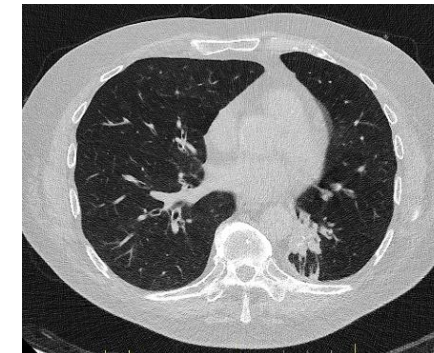
Concept : use more reliable attenuation values for better AC correction.

Measure tissue attenuation from transmission scan (CT).

⚠ CT for AC purposes are low-dose! ☢



CT-based attenuation map is used for attenuation correction of SPECT data



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

Hybrid systems and attenuation correction (AC)

Vacuum tube with cathode (filament heated through an electrical current).

e⁻ are liberated and accelerated by a bias voltage towards the anode (generally, rotating tungsten plate).

Production of continuous Bremsstrahlung radiation + discrete characteristic X-rays (anode material).

Tube current → # of emitted e⁻

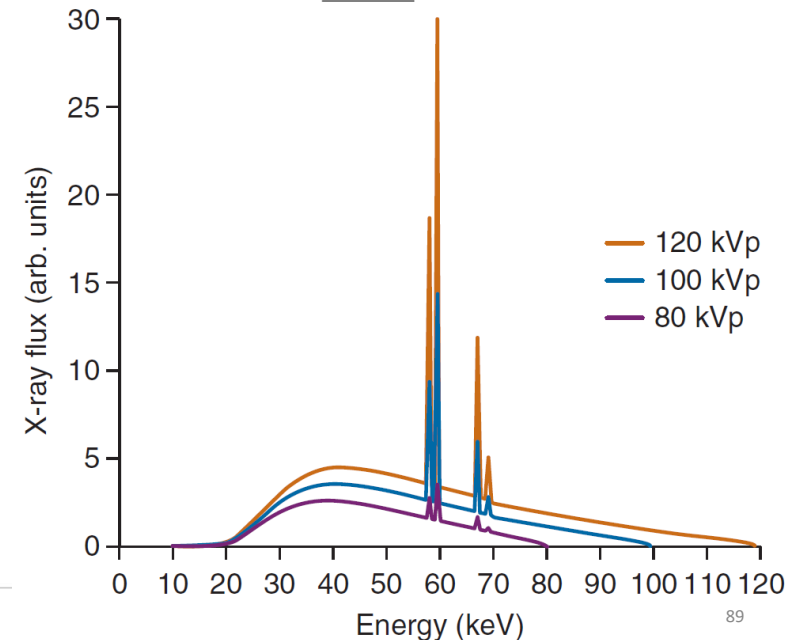
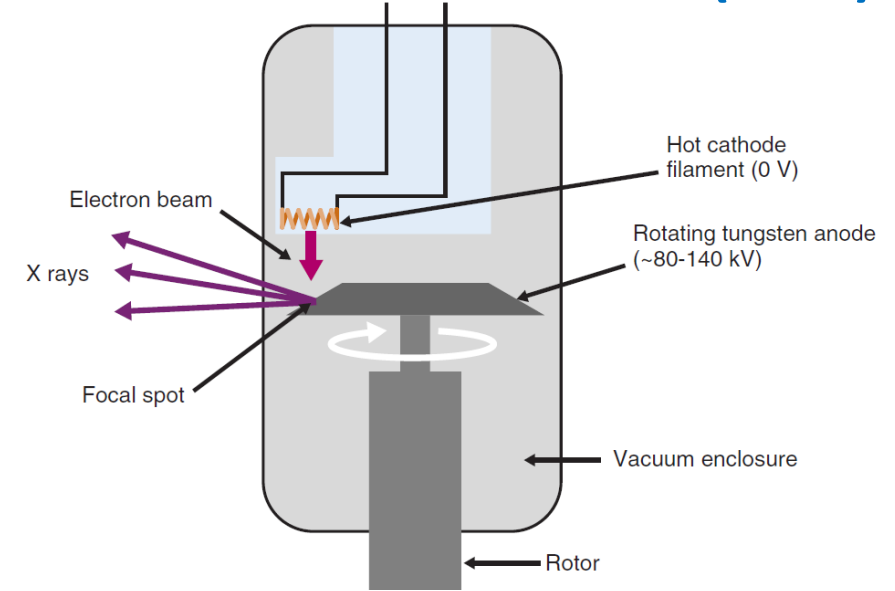
Voltage → energy spectrum of X-rays (max =kVp) and #

X-ray beam collimated and filtered.

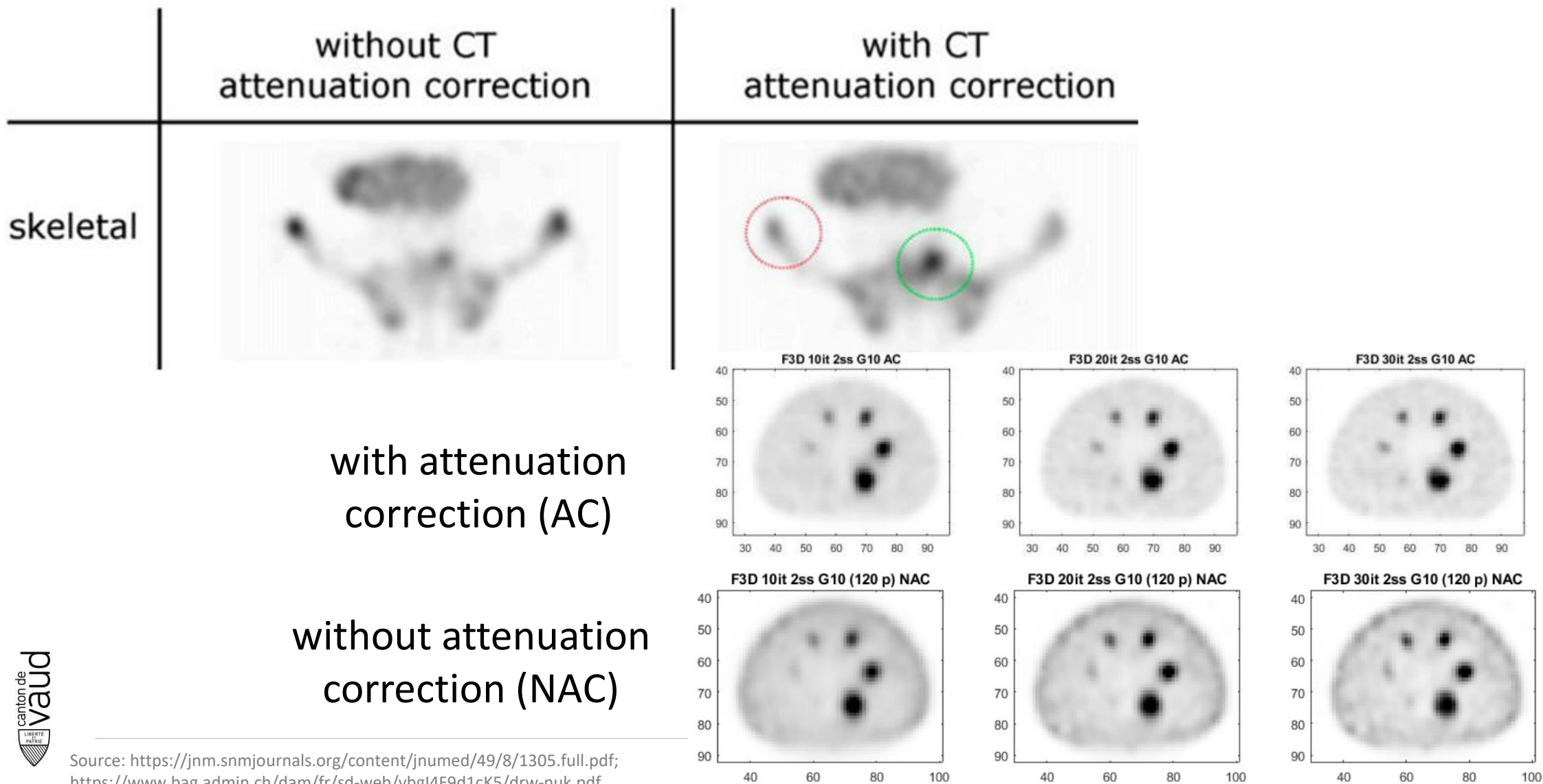
CT provide a map of the attenuation coefficient μ within the body



Conversion of the map of μ acquired with the CT transmission photon energy (e.g. effective energy 90 keV) to the SPECT emission photon energy (e.g. 140 keV for ^{99m}Tc).



Attenuation correction (AC) + diagnostic reference levels (DRL / DRW / NRD)



Attenuation correction (AC) + diagnostic reference levels (DRL / DRW / NRD)

Tableau 1.1 Niveaux de référence diagnostiques lors d'examens de médecine nucléaire chez les adultes

Examen	Nucléide	Produit radiopharmaceutique	NRD (activité)		CT Absorption/Localisation NRD (75 ^e percentile)		Dose effective E ₅₀ due au produit radiopharmaceutique [mSv]
			pour 70 kg [MBq]	par poids [MBq/kg]	CTDI _{vol} [mGy]	DLP [mGy·cm]	
Système osseux	Tc-99m	DPD (Teceos), MDP (Lenoscint), HDP	700	10,0	10 (bassin) 5 (CV) 5 (extr.)	410 (bassin) 190 (CV) 160 (extr)	4,0
Thyroïde	I-123	Iodure	10		4	160	2,2 ¹
	I-131	Iodure	3				72,0 ¹
	Tc-99m	Pertechnétate	75				0,98
	Tc-99m	MIBI (Cardiolite)	170				1,5
Parathyroïde	I-123	Iodure	20		4	160	4,4
	Tc-99m	MIBI (Cardiolite)	550				5,0
Perfusion pulmonaire	Tc-99m	MAA	180		3	110	2,0
Ventilation pulmonaire	Xe-133	Gaz	400				0,072
	Tc-99m	Aérosol (DTPA) ²	1000				0,31 ³
	Tc-99m	Technegas ⁴	500		0,38 ³		

Directive BAG/OFSP :

Diagnostic reference levels for the administered activity and CT for attenuation correction

Quiz

After handling unsealed radioactive sources, workers must perform a screening measurement to verify the absence of significant (accidental) radionuclide intake.

A nuclear medicine technologist worked with the following radionuclides: Ga-68 and I-131.

Which radiation protection instruments would you recommend to perform the screening measurement?



From the Swiss dosimetry ordinance (814.501.43)



17. Ga-68

17.1 Métabolisme

En raison de sa courte durée de vie (période physique de 68 min.), le Ga-68 se désintègre presque entièrement dans le corps avant d'être éliminé. Le Ga-68 inhalé ou ingéré dépose principalement sa dose dans les poumons (inhalation) et/ou dans le tractus gastro-intestinal (ingestion).

17.2 Méthodes de mesure

Mesure de tri (obligatoire)

Mesure du rayonnement direct au niveau de l'estomac/de l'abdomen, au minimum toutes les 4 heures.

Seuil de mesure: 1 $\mu\text{Sv/h}$ au niveau de l'estomac

Mesure d'incorporation

A cause de la courte période physique, une mesure d'incorporation n'est pas possible.

17.3 Intervalles de surveillance T et laps de temps t entre l'événement et la 1^{re} mesure

T_{tri} :	4 heures	T_{mesure} :	—	$t_{\text{événement}}$:	immédiatement
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From the Swiss dosimetry ordinance (814.501.43)



27. I-131

27.1 Métabolisme

L'iode inhalé (classe d'absorption type F) est exhalé à 50 %. L'autre moitié atteint rapidement la circulation sanguine (taux de résorption $f_1 = 1$). De là environ 30 % est résorbé en 1 jour dans la glande thyroïde et 70 % est éliminé par voie urinaire. La période biologique dans la glande thyroïde est de 80 jours. La durée de séjour de l'iode-131 dans la thyroïde est ainsi déterminée par sa période physique de 8 jours.

27.2 Méthodes de mesure

Mesure de tri

Mesure directe de l'activité fixée dans la glande thyroïde avec un moniteur de contamination.

Seuil de mesure: 2000 Bq

Mesure d'incorporation

Mesure à l'aide d'un moniteur thyroïdien de l'activité de I-131 M en Bq.

27.3 Intervalles de surveillance T et laps de temps t entre l'événement et la 1^{re} mesure

T_{tri} :	7 jours	T_{mesure} :	30 jours	t événement:	6–12 h
--------------------	---------	-----------------------	----------	--------------	--------

Result interpretation

when I placed the instrument close to my belly, it displayed «1.1 $\mu\text{Sv/h}$ »



Did you subtract the background?

when I placed the instrument close to my thyroid, it displayed «5.4 counts per second (net)»



How can you interpret this result?

Result interpretation



**Institut de
radiophysique**
Rue du Grand-Pré 1
CH-1007 Lausanne



Kalibrierstelle für Strahlenschutzmessgeräte
Laboratoire d'étalonnage d'instruments de mesure pour la radioprotection
Laboratorio di calibrazione per strumenti di misura per la radioprotezione

Zertifikat Nr.

N° du certificat

Thyr-2021-237

N° del certificato

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Facteur d'étalonnage et valeur limite pour la mesure de tri

Radioisotope	Sensibilité [cps/Bq]	Facteur d'étalonnage [Bq/cps]	Seuil de mesure de tri de l'ordonnance [Bq]	Limite <i>nette</i> pour la mesure de tri [cps]
I-131	0.00072	1387.66	2000	1.4
I-125	0.00282	354.39	1300	3.7
I-123	0.00203	492.71	1400	2.8

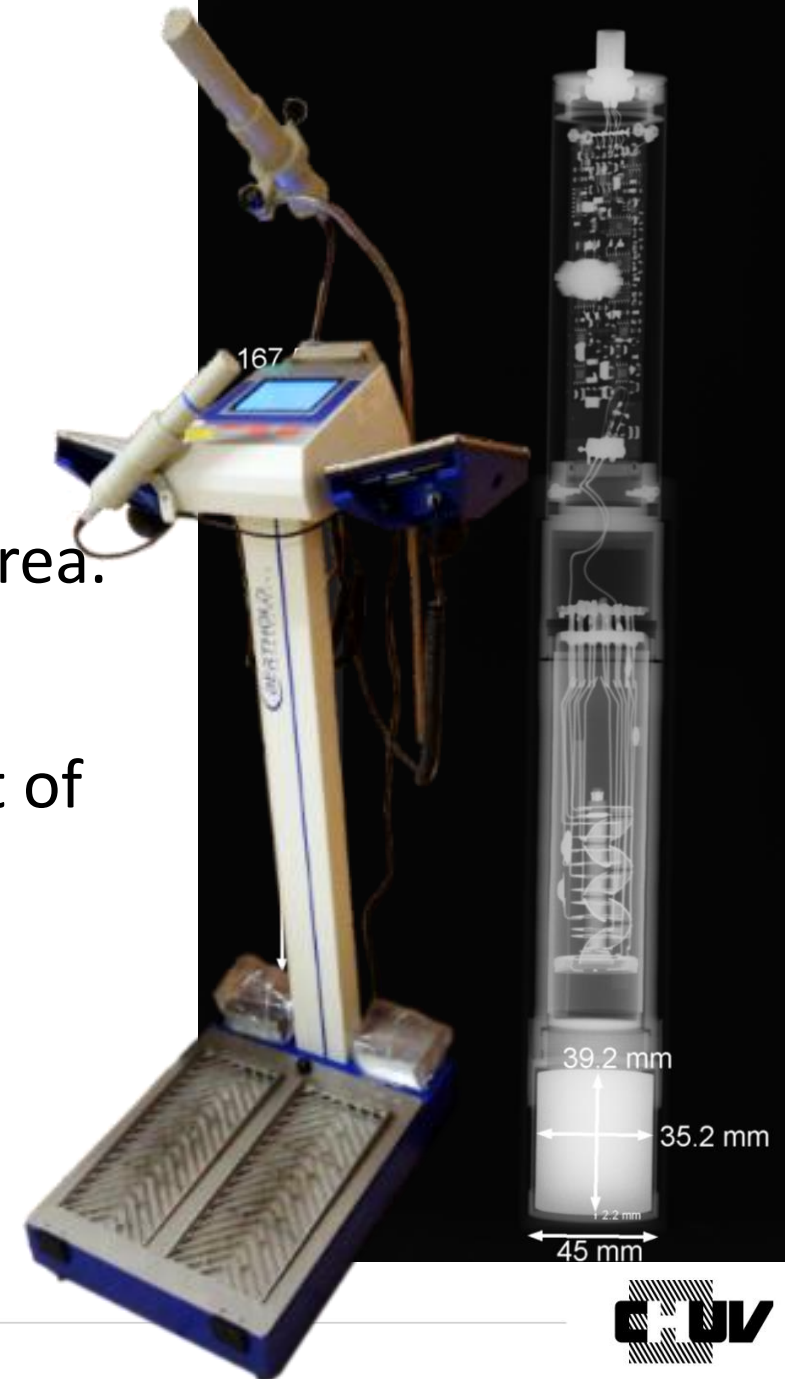
L'incertitude élargie de l'étalonnage est de 12%. Il s'agit de l'incertitude-type de l'étalonnage multipliée par le facteur d'élargissement $k=2$ ce qui, pour une distribution gaussienne, correspond à un niveau de confiance d'environ 95%.

Quiz

The nuclear medicine service recently bought this instrument to check for the absence of external and internal contamination when leaving the controlled area.

You lack technical details on the probe placed in front of the thyroid, thus decide to perform a radiography.

Which kind of detector is it?



Quiz

CERN

The ~~nuclear medicine~~ technologist is contaminated, but he/she handled several unsealed sources today.

How would you proceed?

Did you think about the available instruments @ the workplace?



Quiz

This instrument:

1. Measures the radioactive concentration of a solution (e.g. urine sample)
2. Measures contamination of a surface (e.g. gloves)
3. Measures the dose rate of a sample (e.g. a syringe)
4. Can be used to measure α radiation



Quiz



My passive dosimeter shows 0 mSv at the end of the month

1. It fulfilled its role to protect me from radiation!
2. This can be completely normal
3. My active dosimeter indicated $102 \mu\text{Sv} = 0.1 \text{ mSv}$. I stop using the passive dosimeter because, obviously, it doesn't work



- [Art. 32 Arrondissement des valeurs de dose](#)

Ordinance on dosimetry

¹ Les services de dosimétrie individuelle arrondissent les valeurs de mesure en mSv à un chiffre après la virgule, après soustraction du bruit de fond.

² Par dérogation, dans le domaine des faibles doses ($< 1 \text{ mSv}$), les valeurs de mesure des dosimètres individuels pour le rayonnement photonique inférieures à 0,075 mSv sont arrondies à 0 et celles supérieures ou égales à 0,075 mSv, à 0,1 mSv.

Source: [https://simpsons.fandom.com/fr/wiki/Radioactive_Man_\(personnage\)](https://simpsons.fandom.com/fr/wiki/Radioactive_Man_(personnage)), adapted from :

https://fr.mirion.com/hubfs/1_Microsites/French%20Website/Nuclear/Spec%20Sheets/Protection%20du%20Personnel/DMC%203000%20Beta%20Module_Spec%20Sheet_SPC-232-FR-A.pdf?hsLang=fr

Quiz



A surface contamination monitor

1. Typically identifies the radionuclide responsible for the contamination
2. Usually measures in counts per second
3. Can never measure α radiation
4. Is always unusable in a nuclear medicine department
(Background too high \rightarrow impossible to measure a contaminated surface)

Quiz

A surface contamination monitor

1. Typically used for the detection of contamination
2. Usually not used for the detection of contamination
3. Can never be used for the detection of contamination
4. Is always used for the detection of contamination (Background radiation is not measured)



wipe test

is used for the

detection of contamination

on a contaminated surface)



Summary

- Explain the main **components** of a **gamma-camera/SPECT device** and its **functioning**.
- Explain the **working principle** of different **dosimetric devices** (activimeter, dose rate meter, spectrometer, contamination monitor).
- Identify the different fields of **clinical** and **radiological protection applications**.