RADIATION DETECTORS AND MEDICAL IMAGING

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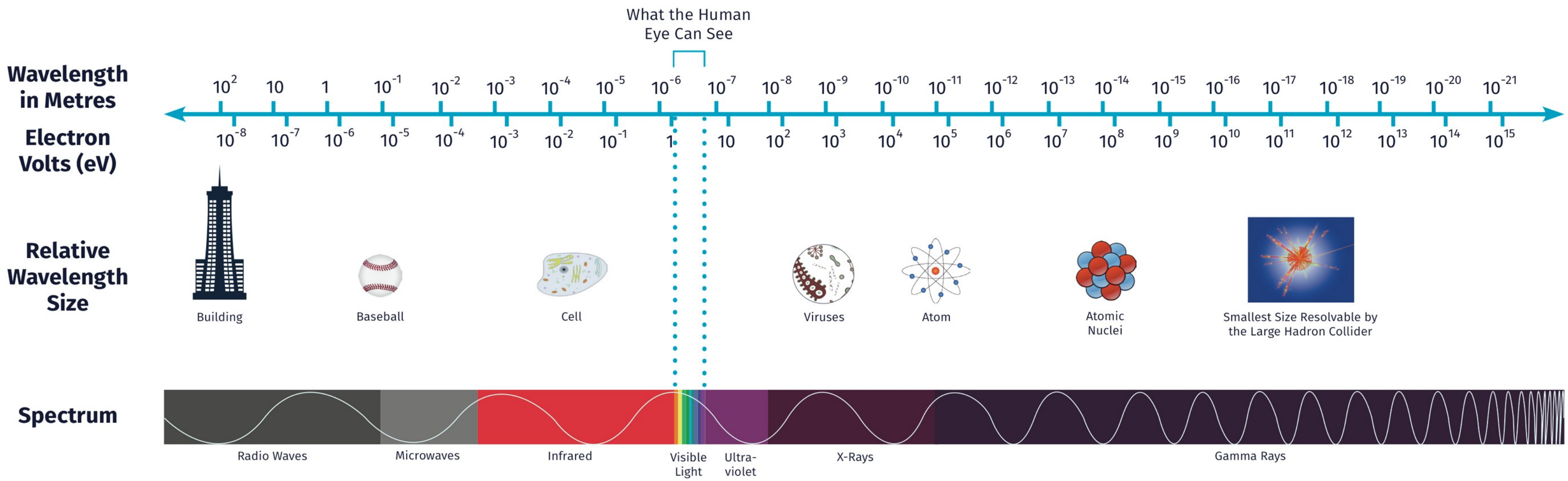
Medical imaging

WHO definition:

Medical imaging encompasses different imaging modalities and processes to image the human body for diagnostic and treatment purposes

In this lecture we will mostly talk about X-ray and nuclear imaging (ultrasound and MRI are not covered)

Electromagnetic spectrum



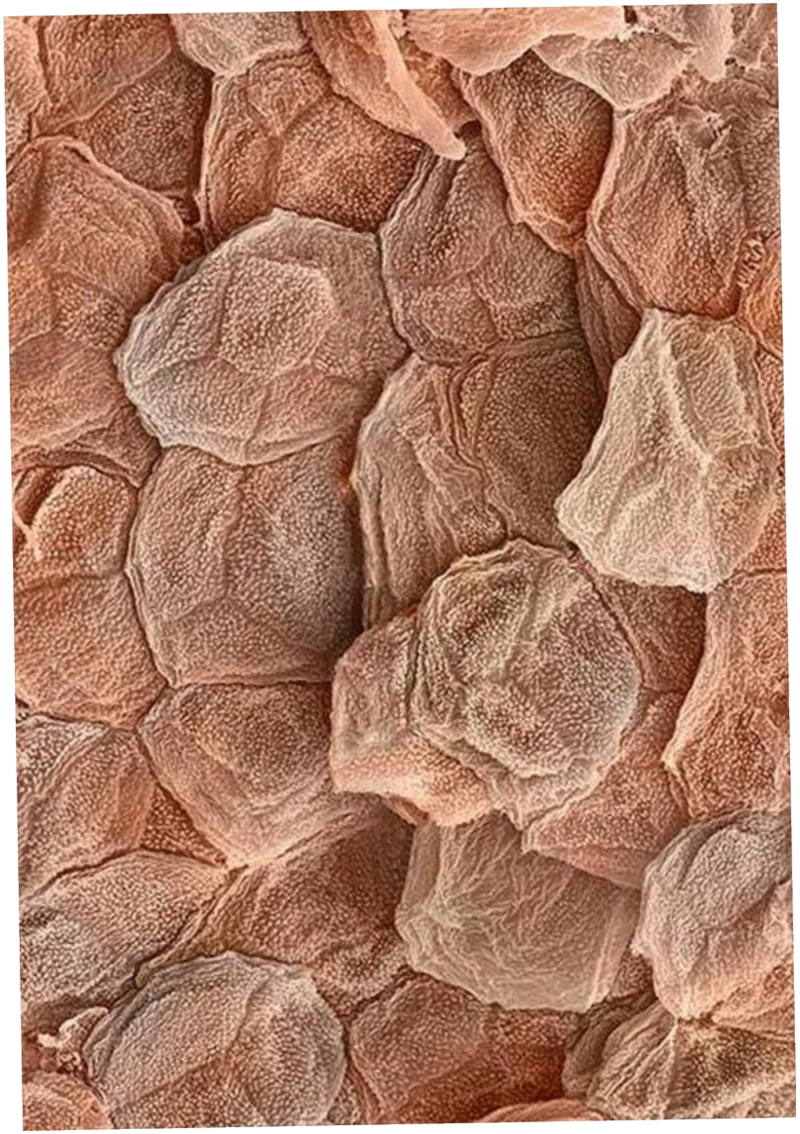
$$E = \hbar\omega = h\nu = \frac{hc}{\lambda}$$

1 eV = 1.6021766208(98)×10^{-19} Joule

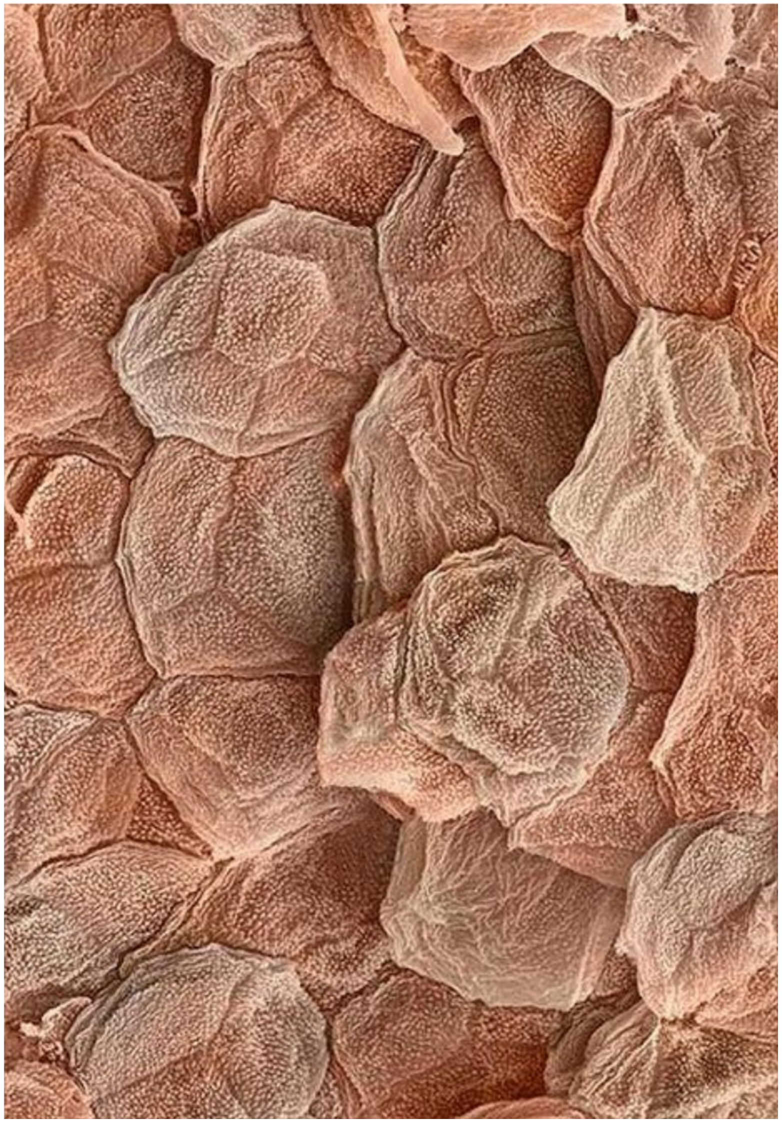
1 keV ~ 1 nm
100 keV ~ 0.01 nm
1 MeV ~ 0.001 nm

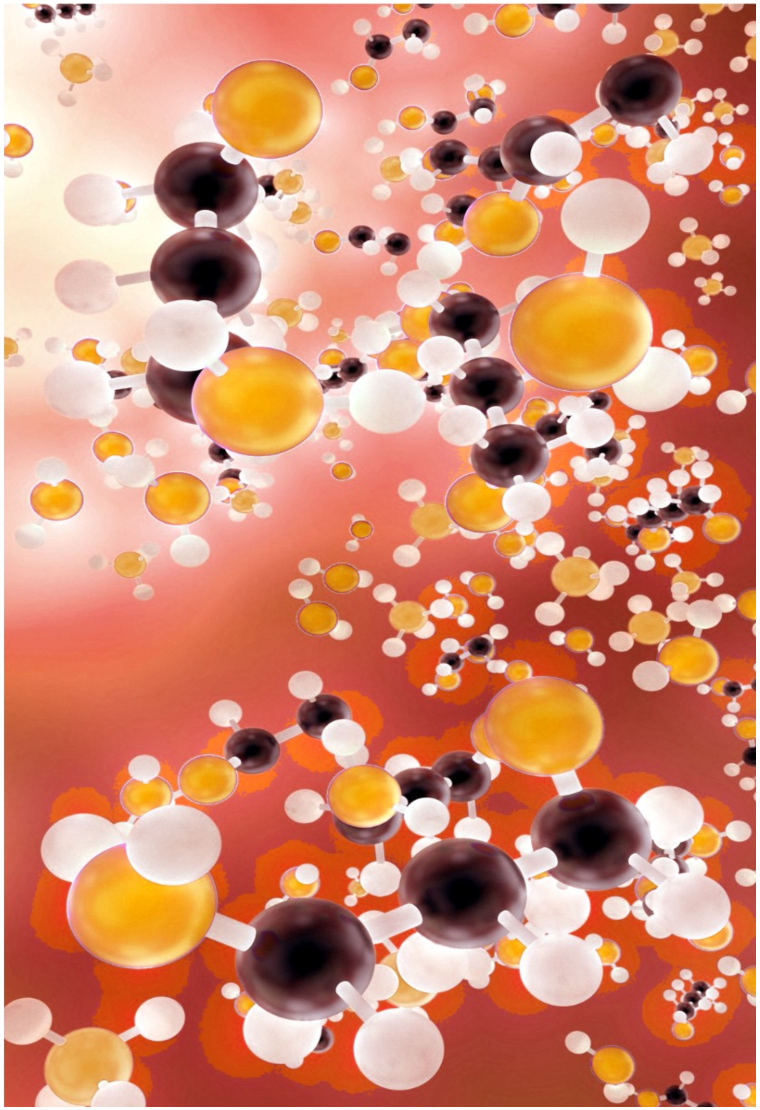
Can rather 'feel' atoms, not molecules!

Effect of detector resolution



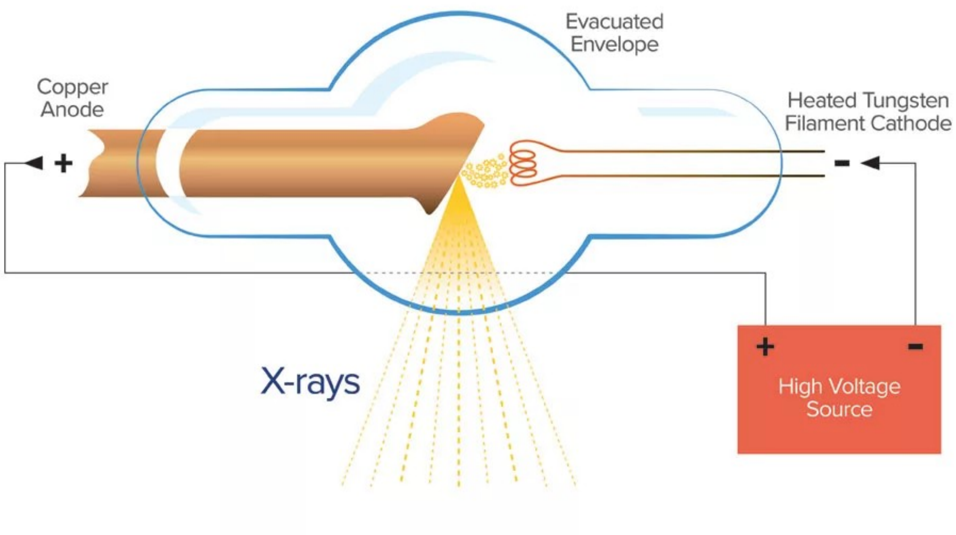
Effect of photon energy (= wavelength)



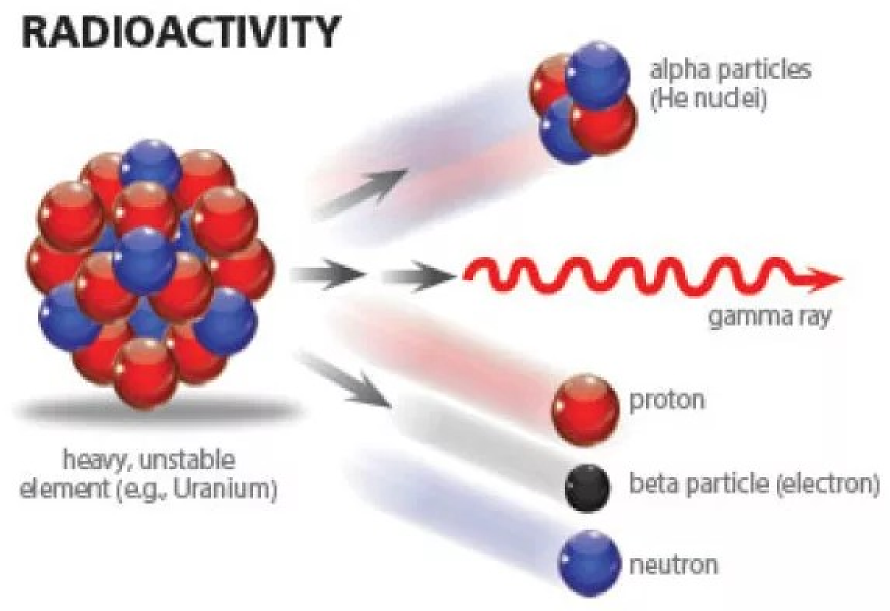


Sources of energetic EM waves

X-rays

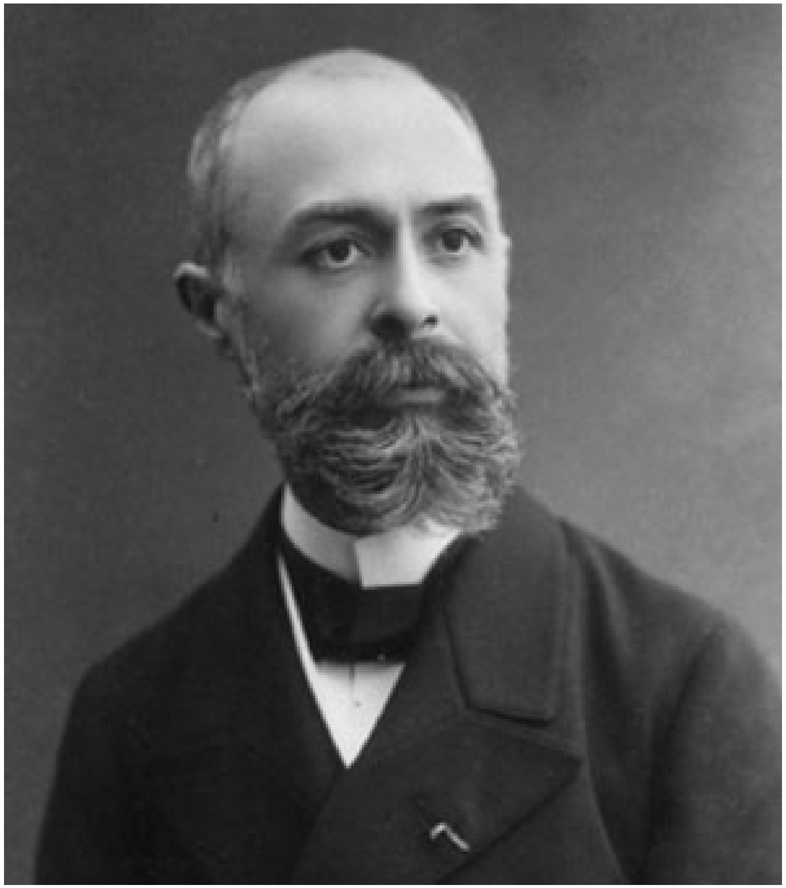


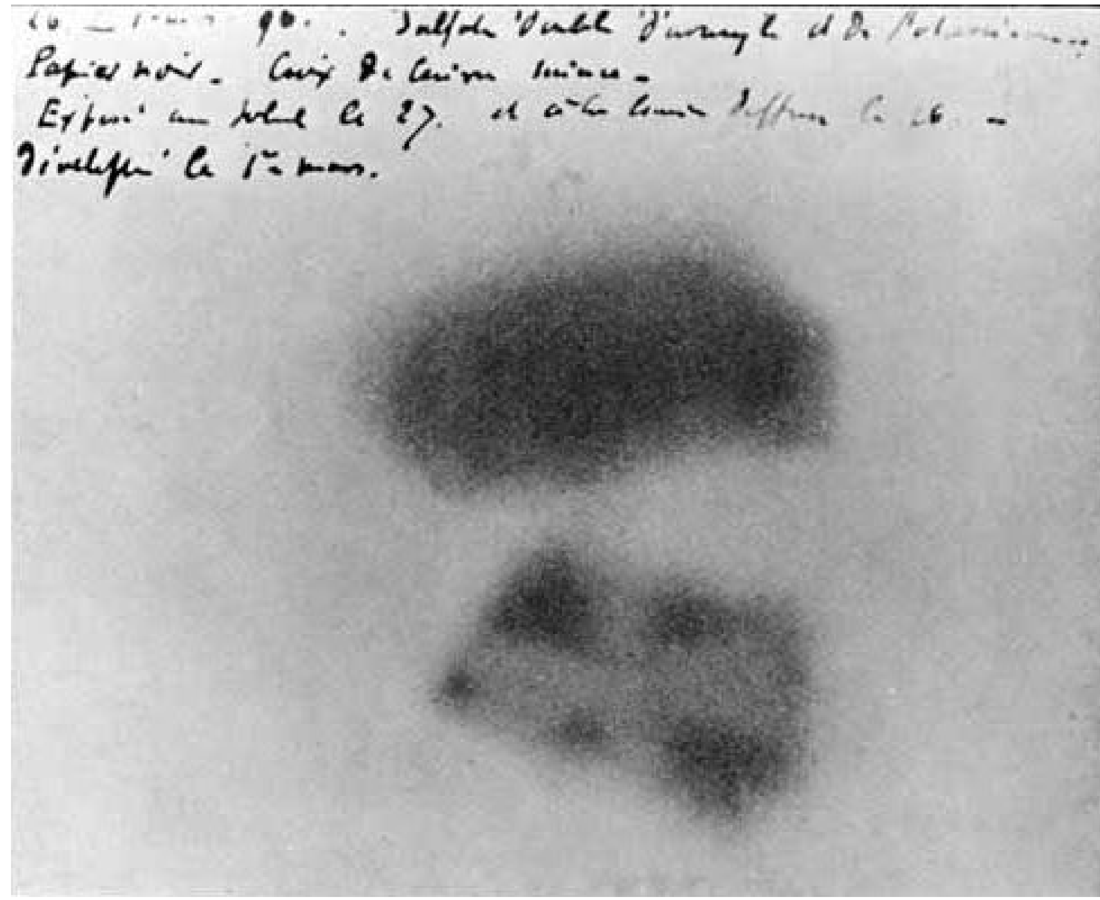
γ rays



The same physical essence
Different origin and slightly different energies


PHYSIK. INSTITUT
DER UNIVERSITÄT

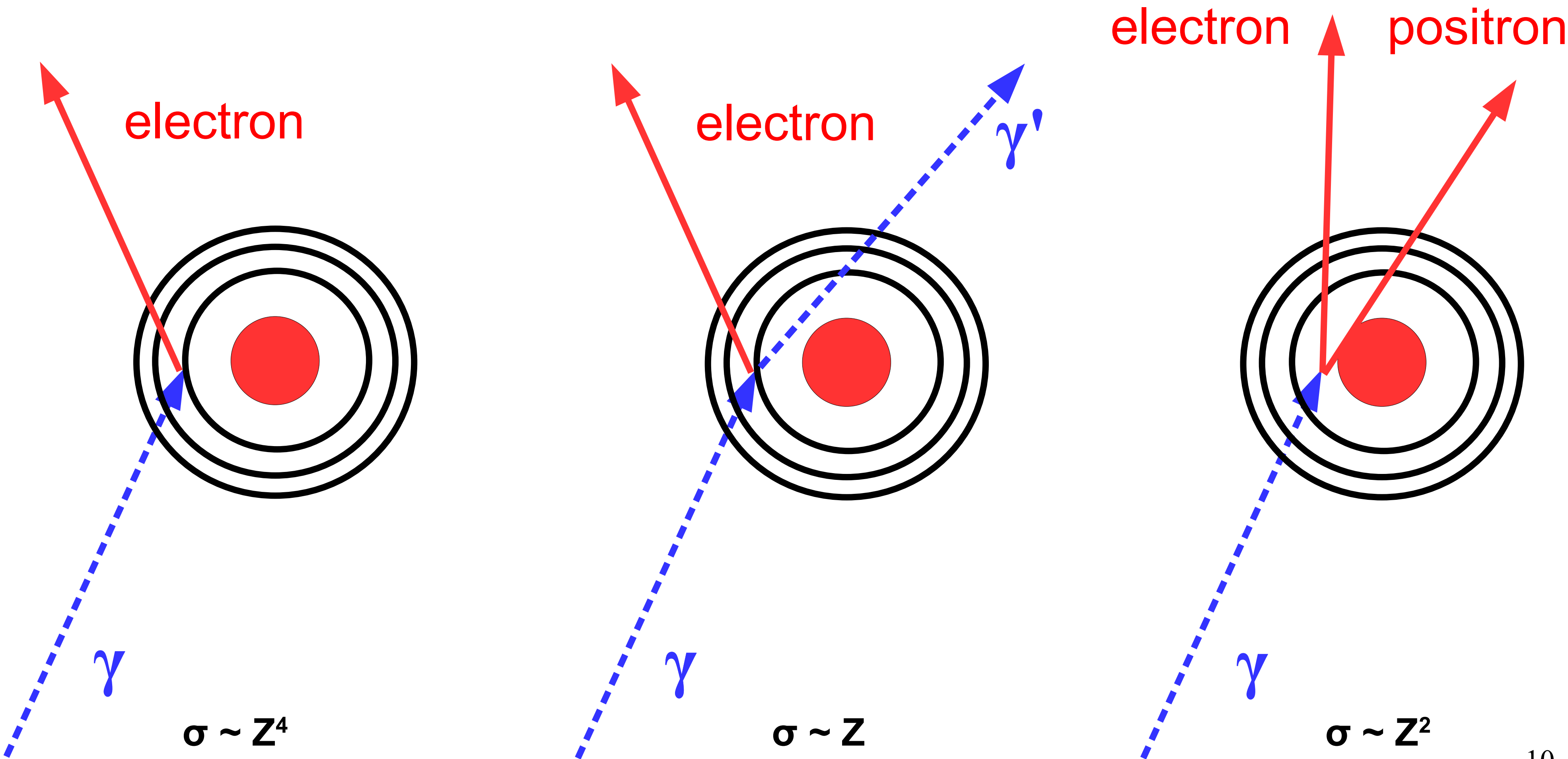




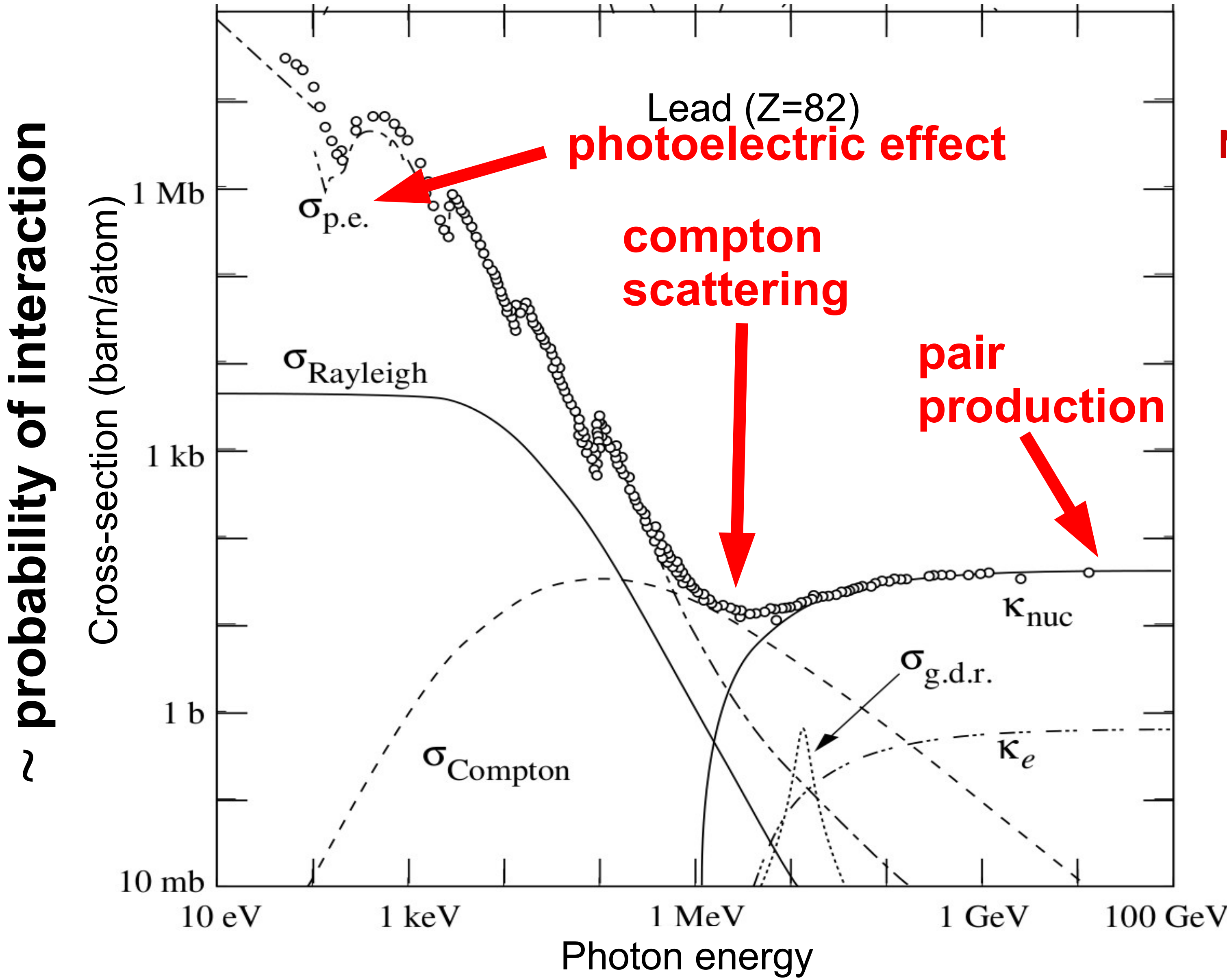
Detectors

Photon interaction with matter

- Photoelectric effect
- Compton scattering
- Pair production



Probability of interaction with matter

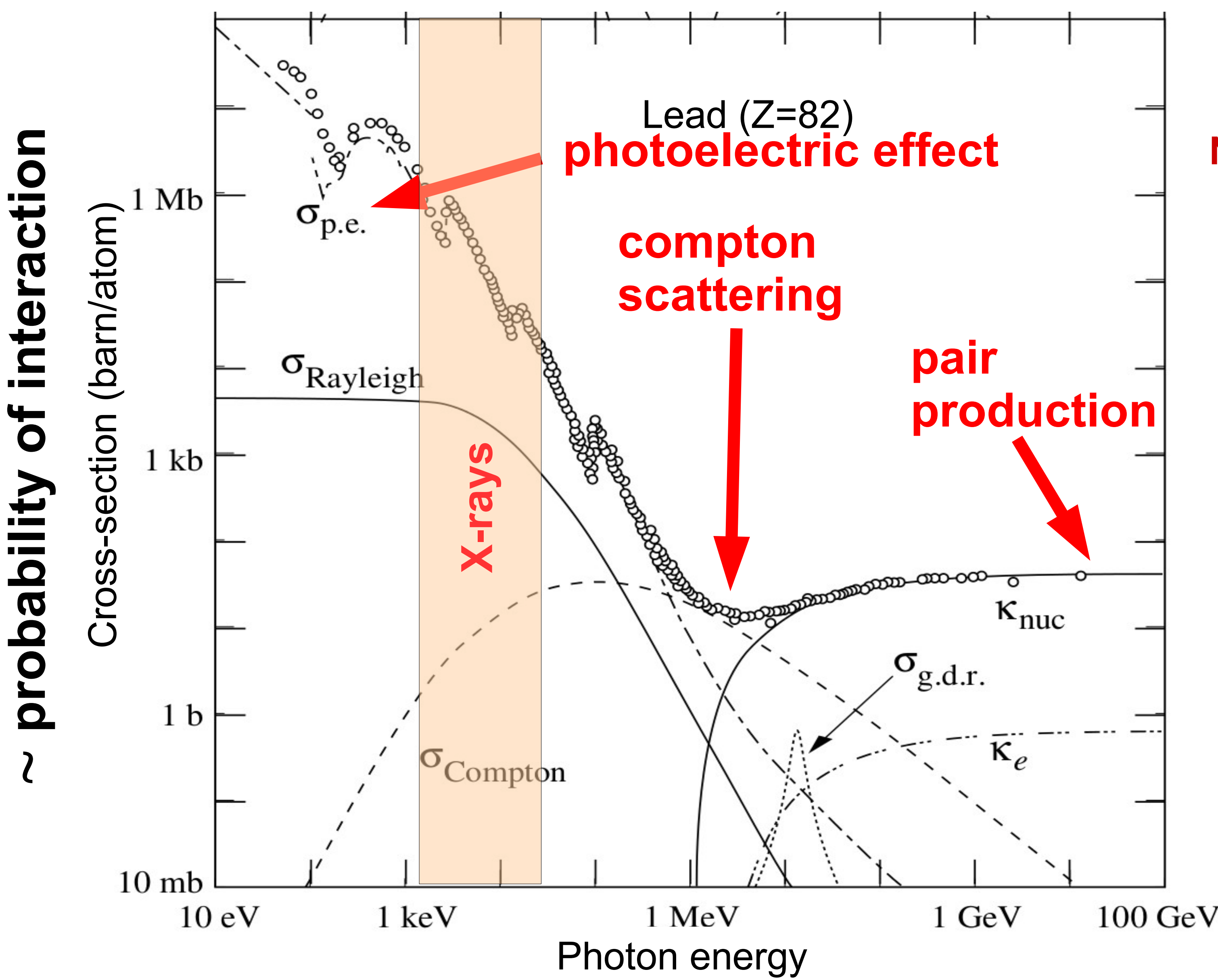


NB: log scale!

PDG2012

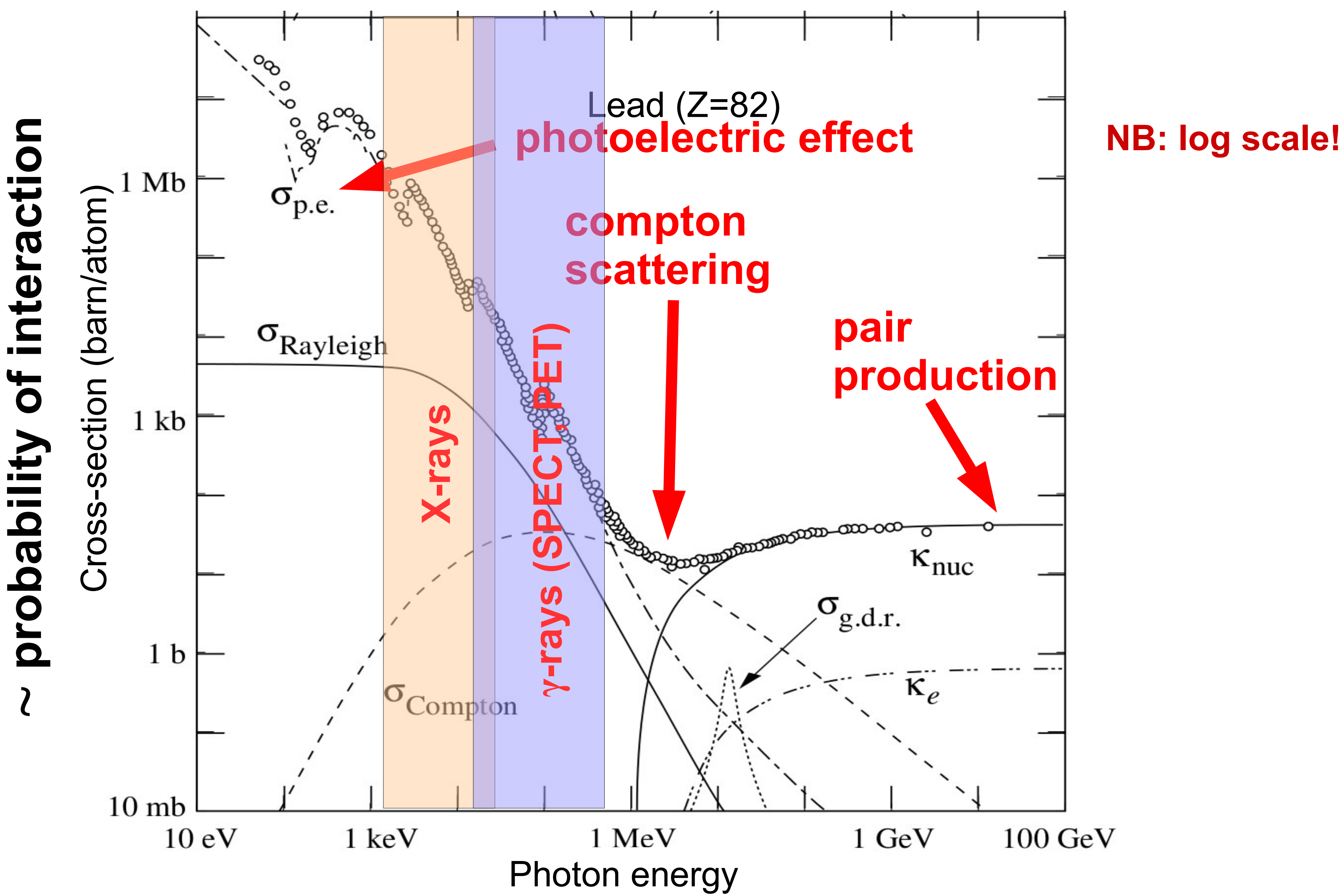
Probability of interaction with matter

NB: log scale!



PDG2012

Probability of interaction with matter



PDG2012

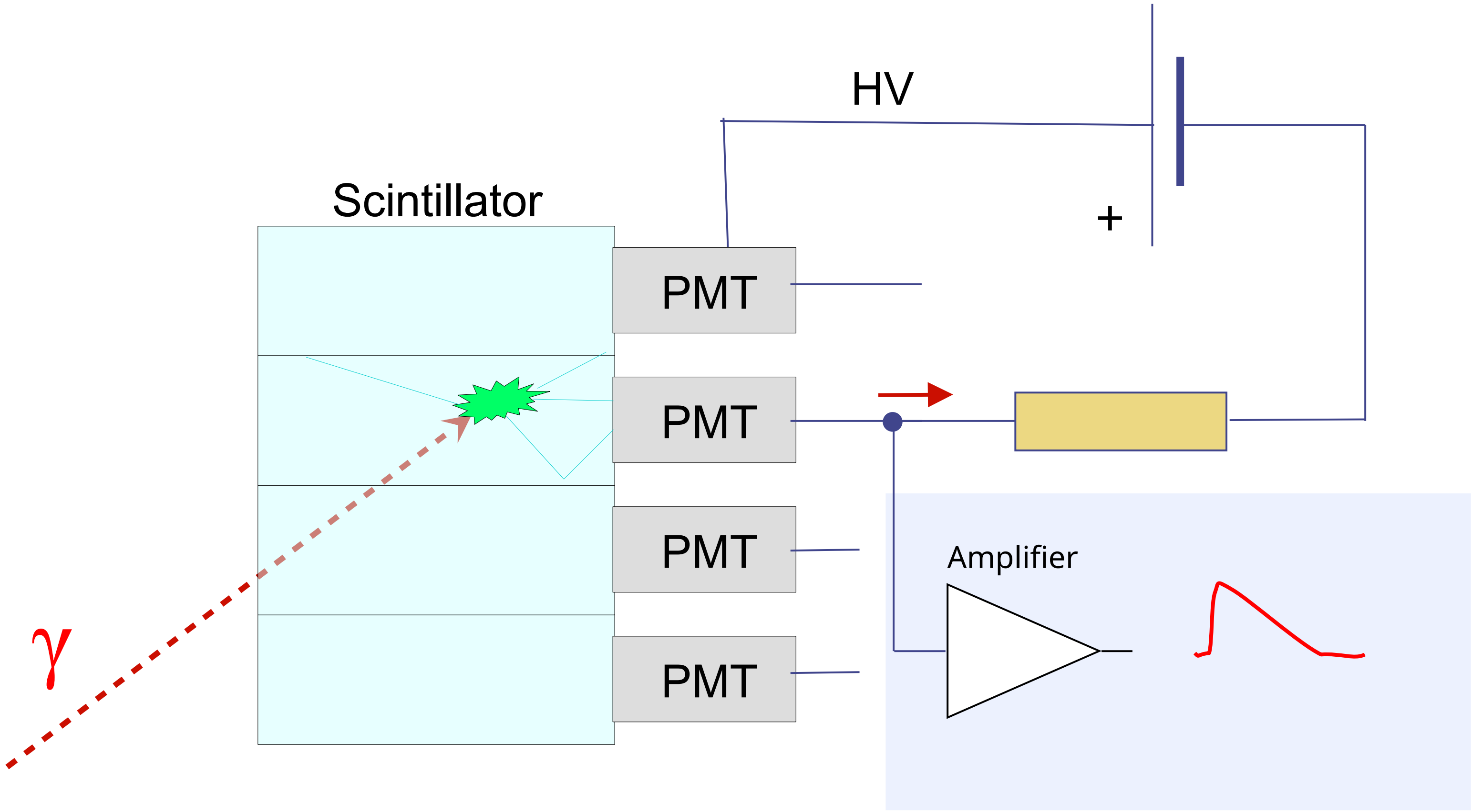
Algorithm of a photon detection

- Photon interacts with the material of the detector and produces electron or electron-positron pair
- Electrons and positrons lose their energy, which is deposited and converted to
  - a darkening of a photofilm or a flash in a fluorescent screen
  - to light (scintillators)
  - to electric charge by ionization (gaseous and solid state detectors)
- Light is collected and registered by a very sensitive photodetector (PMT or SiPM or CCD) producing an electric signal
- Electric charge moves (if electric field is applied) and induces a current pulse in a readout chain

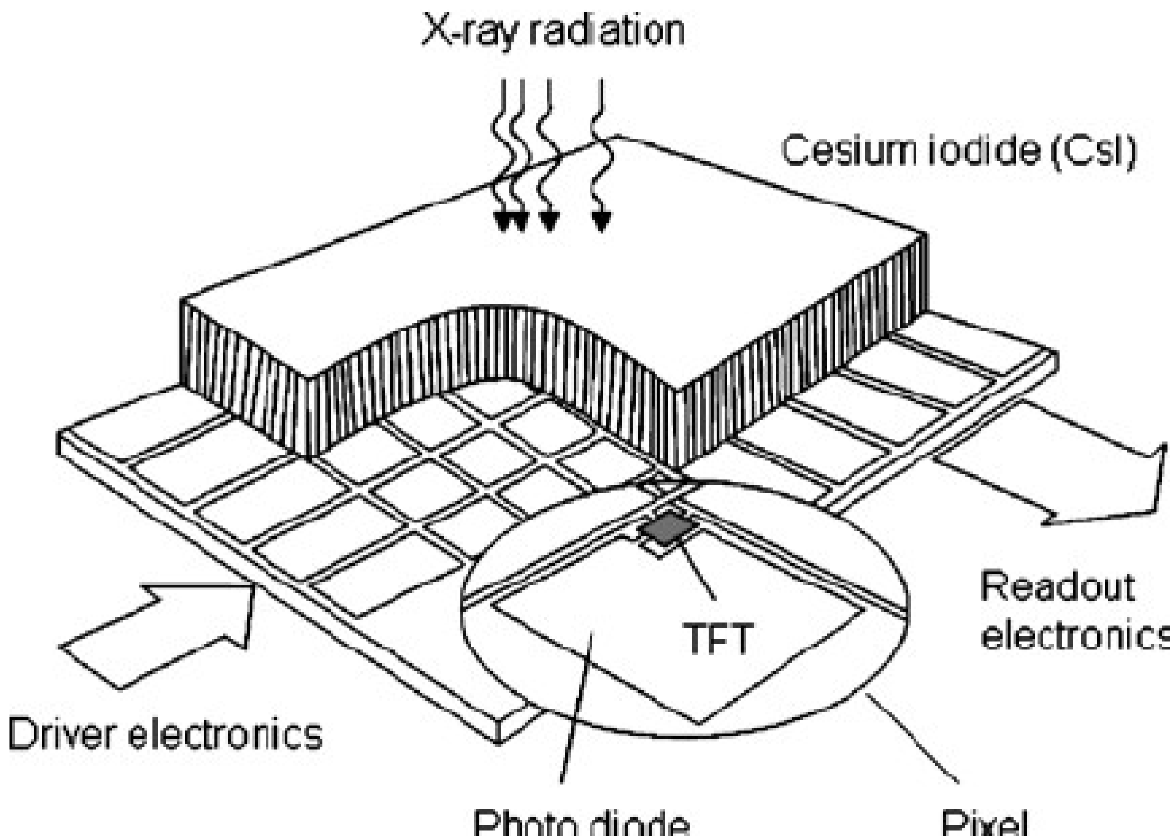
Key propetries of an imaging detector

- Efficiency (determines the patient's radiation dose)
- Resolution
  - spatial
  - (energy)
  - (time)
- Response time
- Noise and uniformity

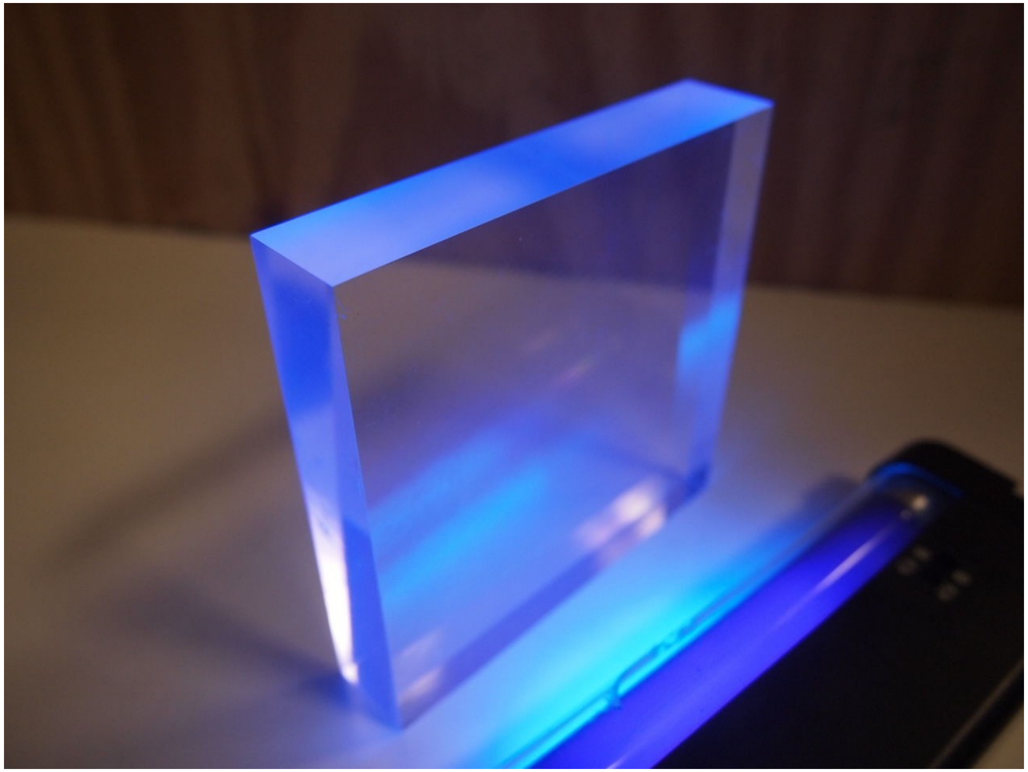
Scintillator detector



PMTs are often replaced by a photodiodes



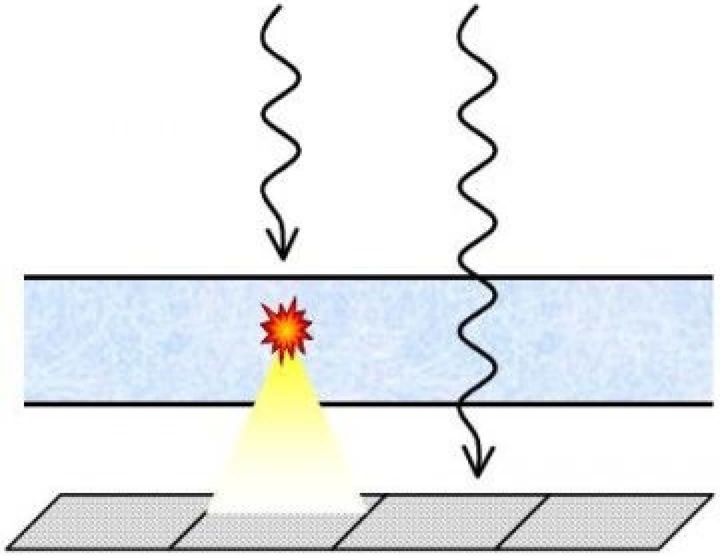
F. Reiser, Maximilian & Becker, C.R. & Nikolaou, Konstantin & Glazer, Gary. (2009). Multislice CT. 10.1007/978-3-540-33125-4.



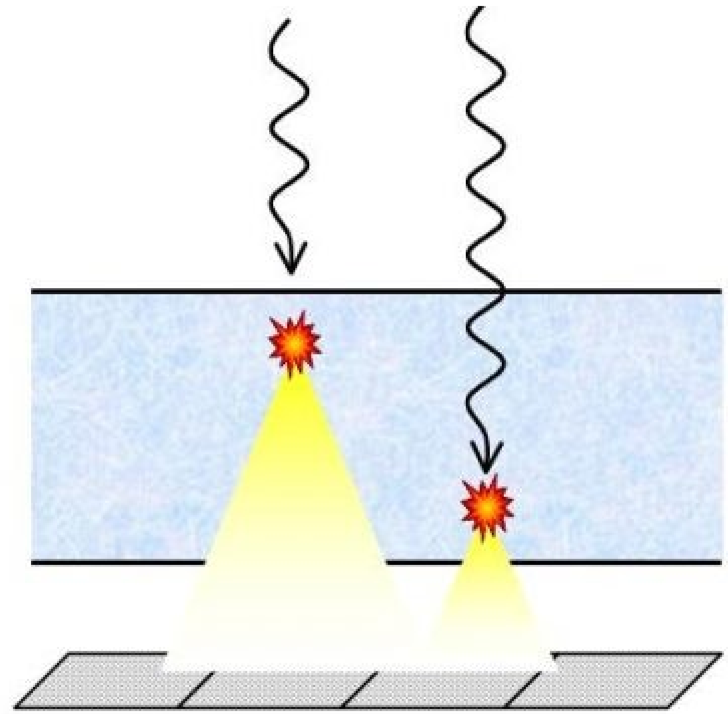
NaI, CsI, BGO, $CdWO_4$, Gd_2O_2S (or UFC)

Properties of scintillator detectors

- Compact and stable
- Long afterglow
- High X-ray detection efficiency (~90%)
- Difficult to obtain spatial resolution < 1 mm

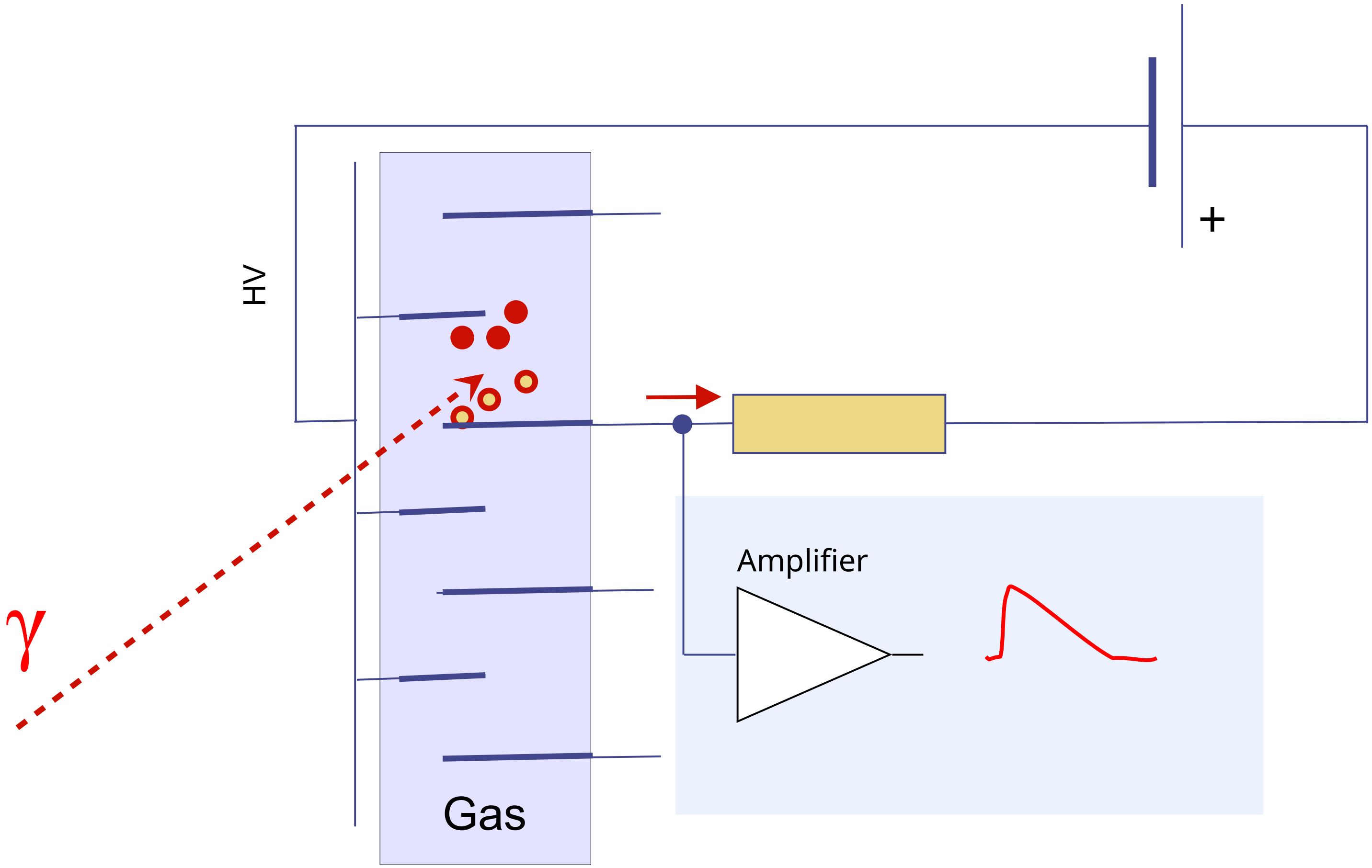


Thin Scintillator:
- high resolution
- low sensitivity



Thick Scintillator:
- low resolution
- high sensitivity

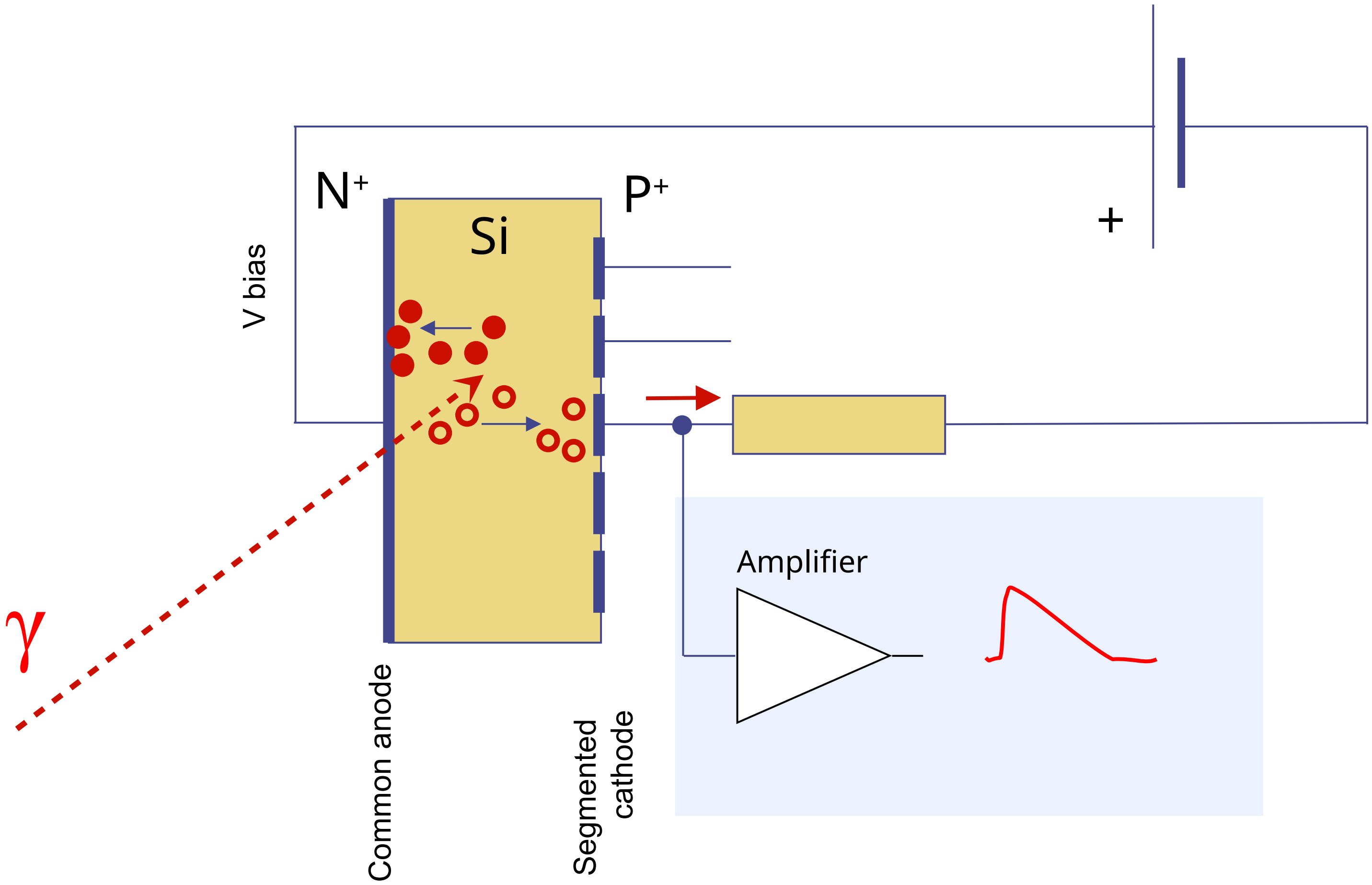
Gaseous detector



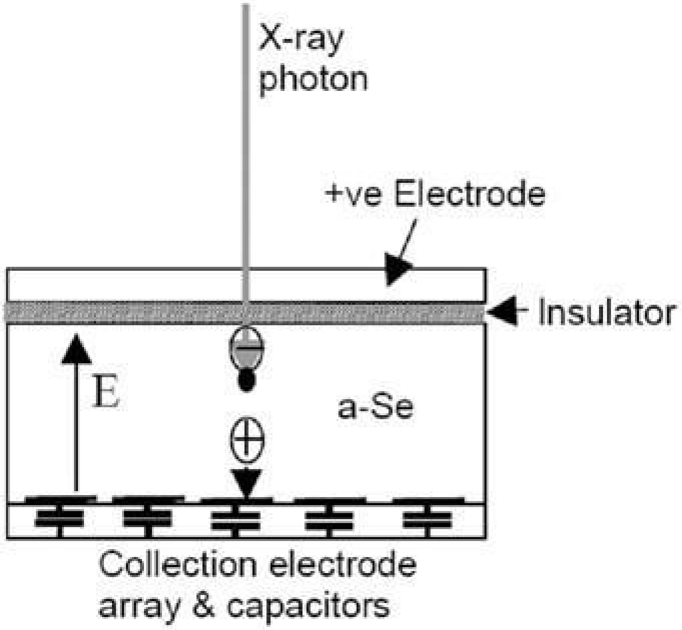
Properties of gaseous detectors

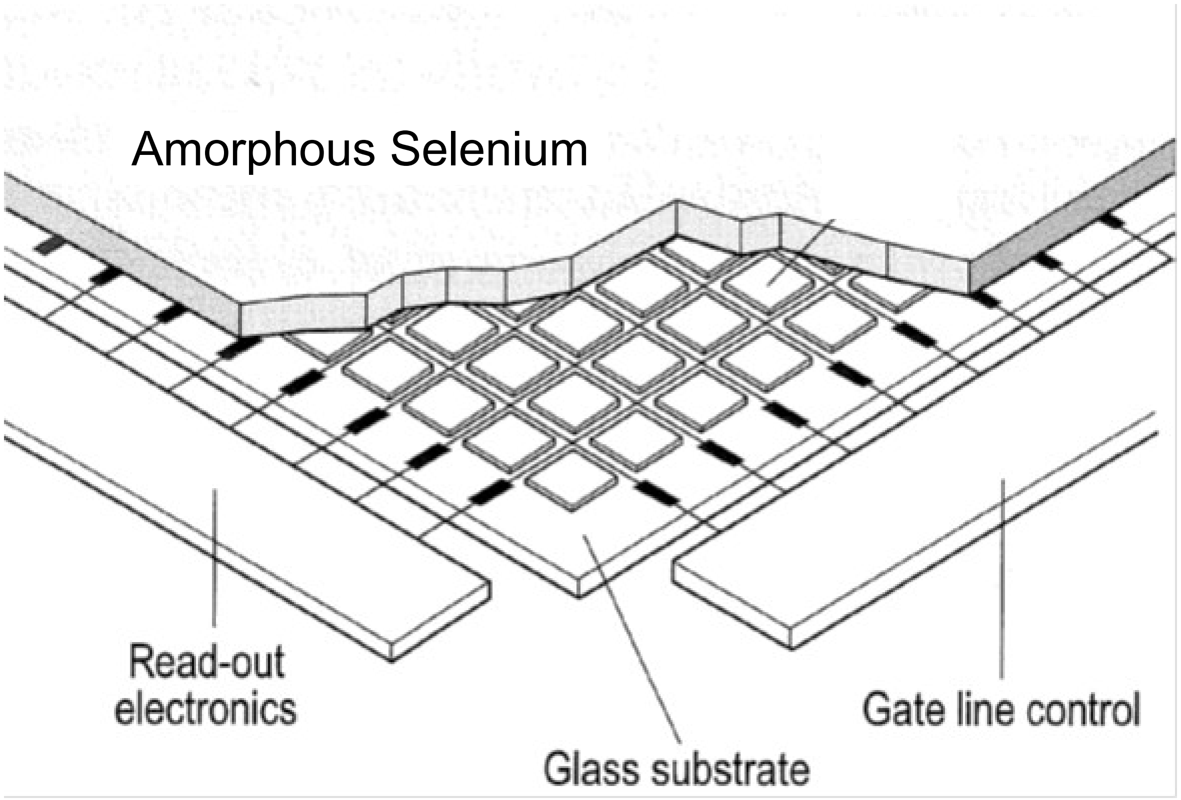
- Excellent stability and uniformity
- Fast response time, no afterglow
- Low X-ray detection efficiency (<50% even with high-Z gas like Xenon and high pressure)
- Difficult to make 2D detectors

Semiconductor detector



Example: a-Se flat panel





Properties of semiconductor detectors

- X-ray detection efficiency is rather low (even for high-Z semiconductors like CdTe, CZT, GaAs, Se)
- Excellent spatial resolution
- Difficult to make large area detectors

Structural imaging vs. Functional imaging

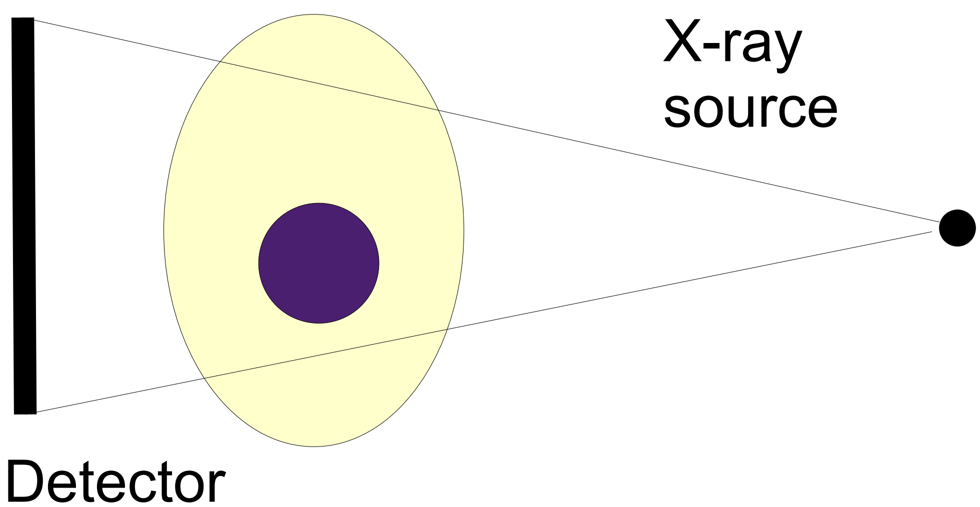
- X-rays allow to obtain images of interior structures of human body, invisible otherwise: shape, density
  → **structural (or anatomic) imaging**
  - good spatial resolution
  - functional change occurs earlier than structural changes
- One can deliver radioactive gamma-sources inside the body and use them as labels to trace their distribution
  → **functional imaging**
  - functional change occurs earlier than structural changes
  - only functional change occurs in some pathologies
  - lacking spatial details

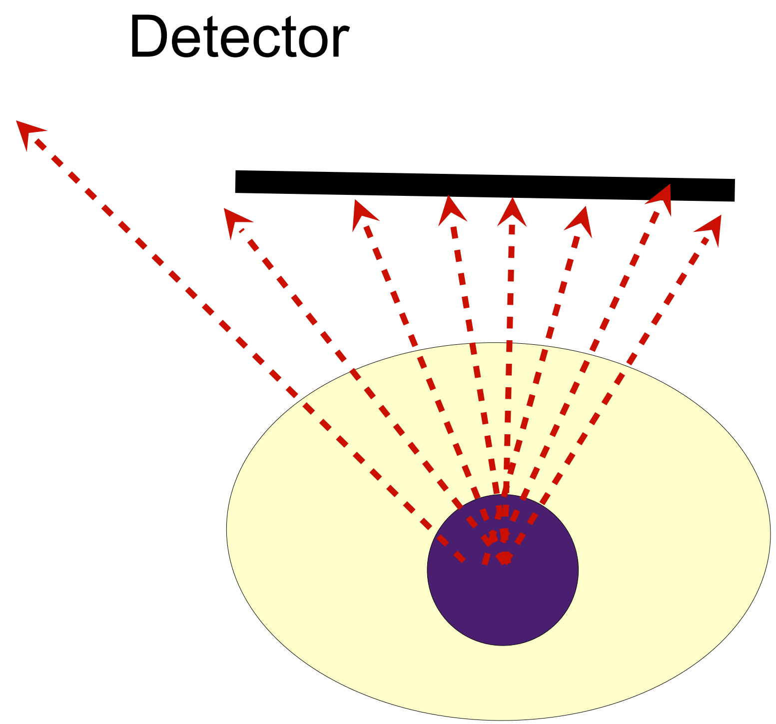
Functional imaging

- Metabolism
- Fluid diffusion
- Blood perfusion

- Cancer diagnostics and metastase location
- Myocardial perfusion
- Stroke assessment
- ....

2D imaging



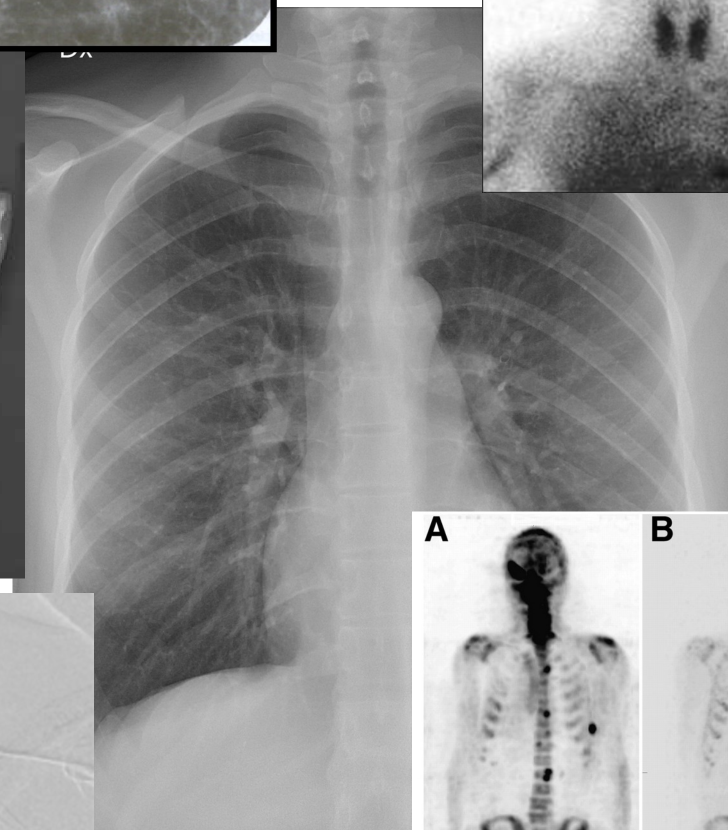


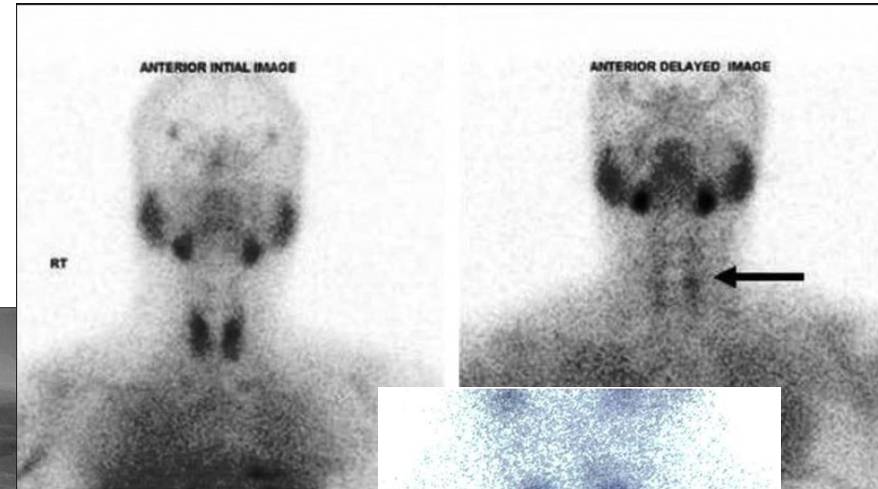
Structural

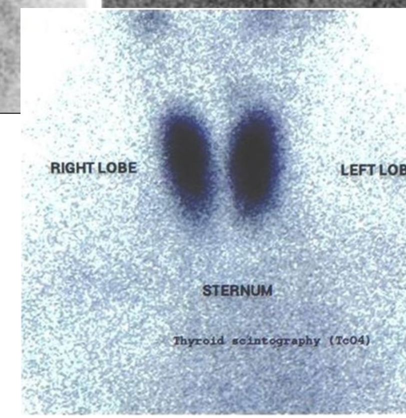
- Radiography
- Fluoroscopy
- Mammography
- Angiography
- ...

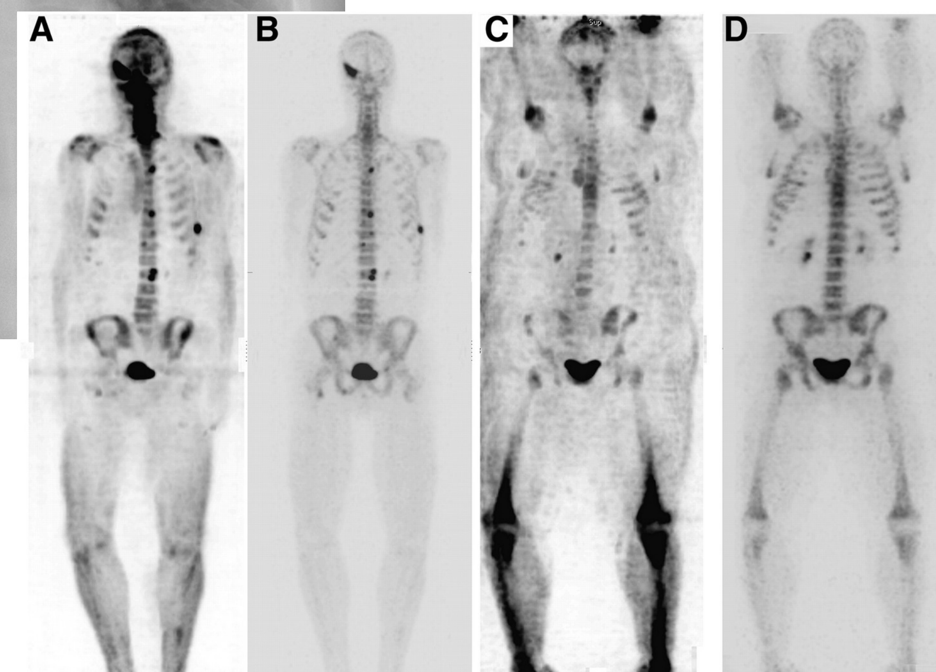
Functional

- Scintigraphy




ANTERIOR INTIAL IMAGE
ANTERIOR DELAYED IMAGE
RT


RIGHT LOBE
LEFT LOBE
STERNUM
Thyroid scintography (TcO4)

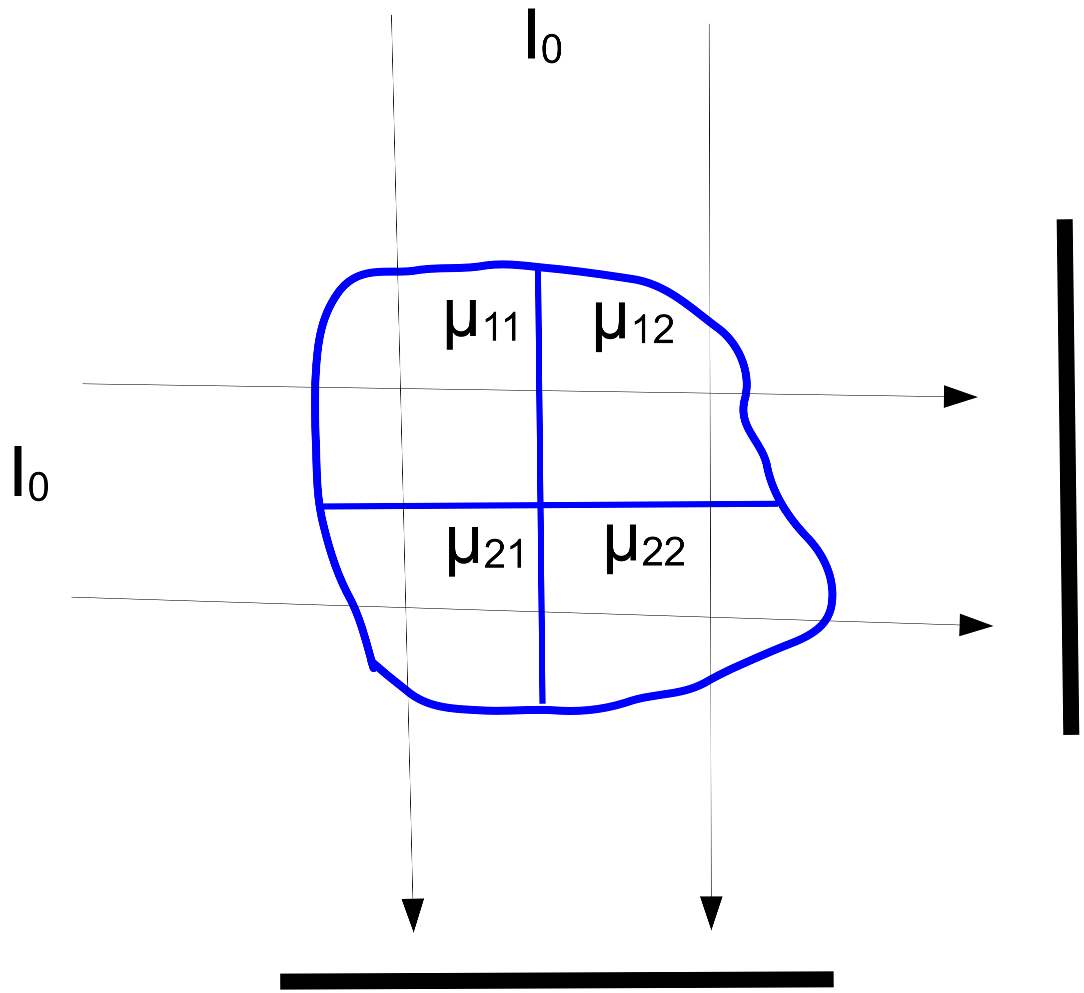

A
B
C
D

Tomography: 2D → 3D

- 2D imaging is simple and quick, but ...
- Two key problems of 2D imaging:
    - superposition of structure shadows
    - image contrast is determined by the full body thickness

Solution: tomography

$$I = I_0 \exp(-\mu x)$$

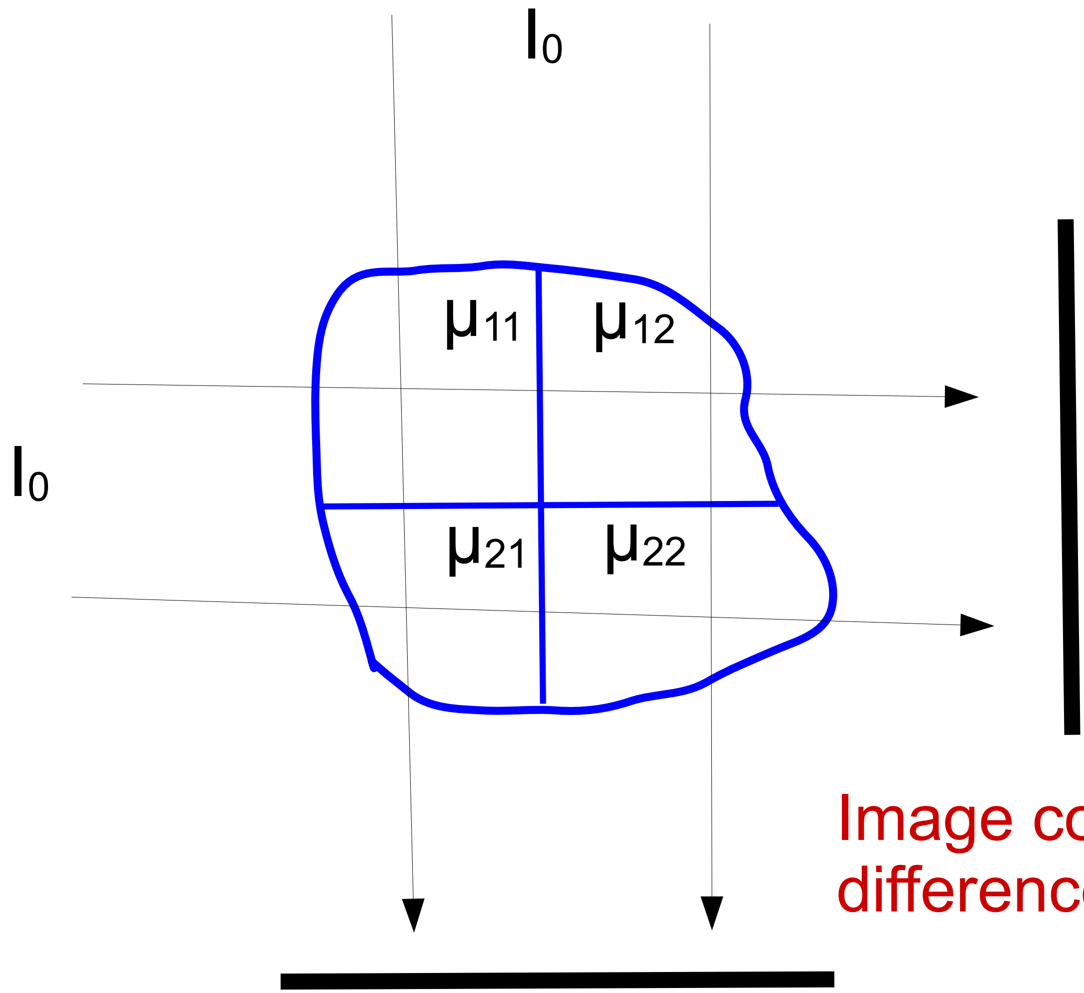


$$I_1 = I_0 \exp(-\mu_{11}x - \mu_{12}x)$$

$$I_2 = I_0 \exp(-\mu_{21}x - \mu_{22}x)$$

$$I_3 = I_0 \exp(-\mu_{11}x - \mu_{21}x) \qquad I_4 = I_0 \exp(-\mu_{12}x - \mu_{22}x)$$

$I = I_0 \exp(-\mu x)$

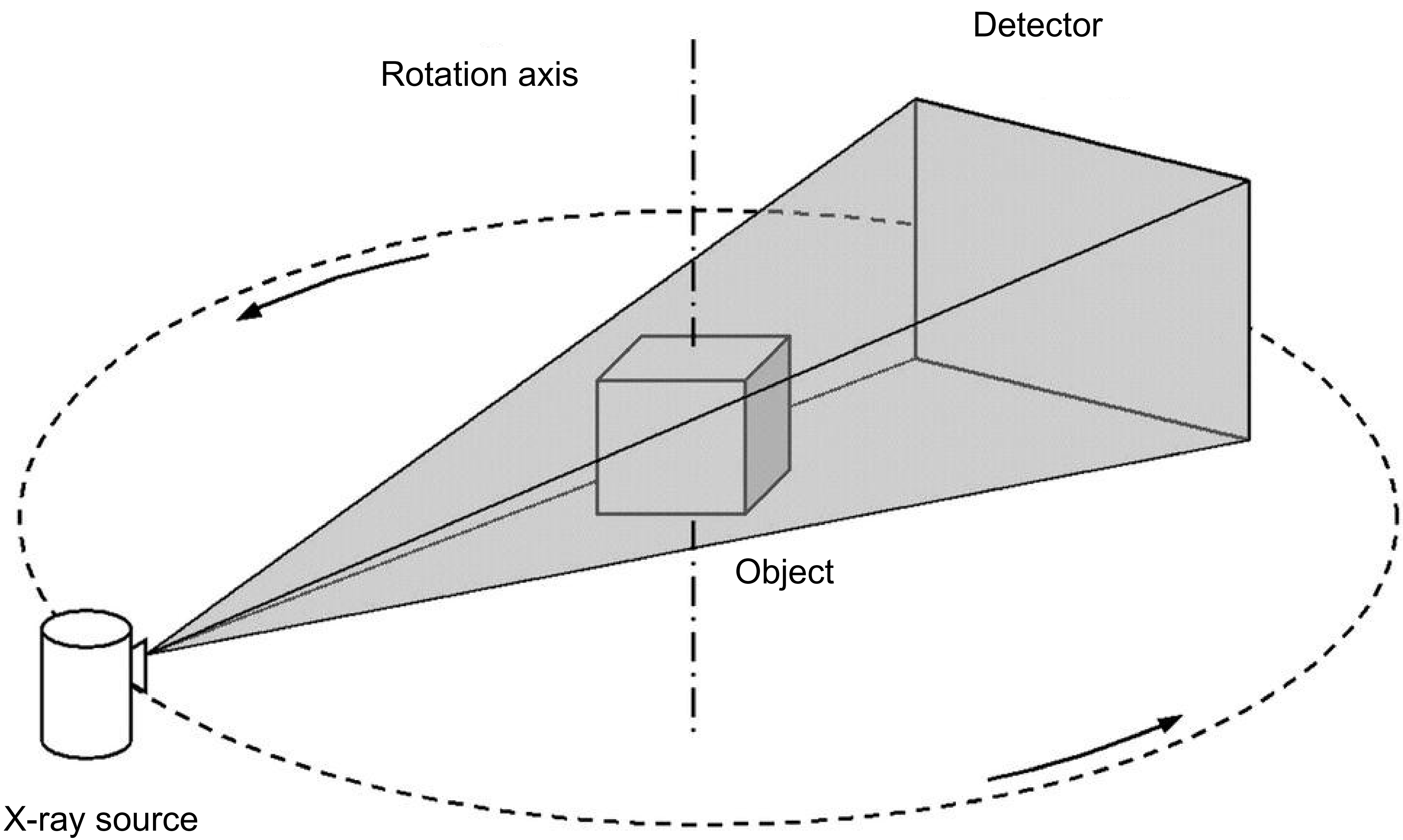


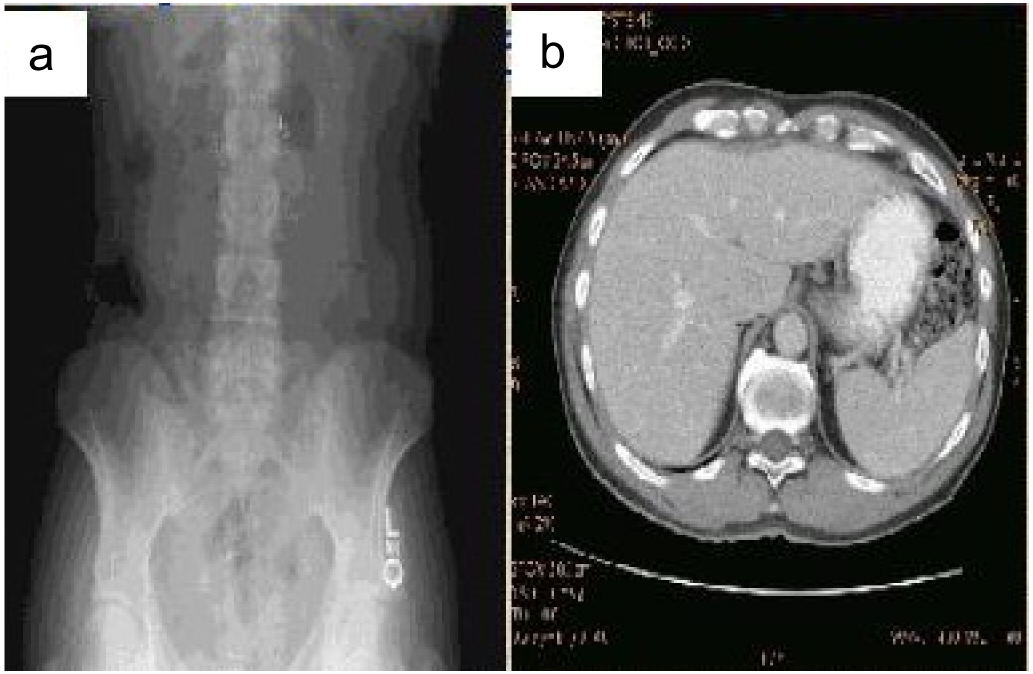
$I_1 = I_0 \exp(-\mu_{11}x-\mu_{12}x)$

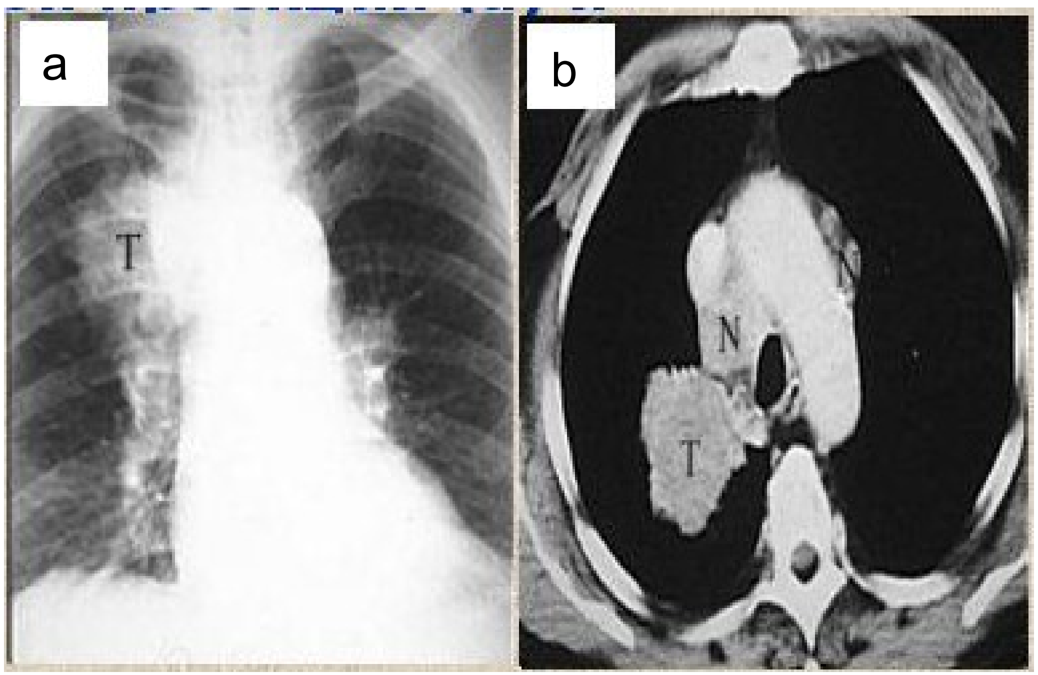
$I_2 = I_0 \exp(-\mu_{21}x-\mu_{22}x)$

Image contrast is determined by the difference of local absorption coefficients

$I_4 = I_0 \exp(-\mu_{11}x-\mu_{21}x)$ $I_4 = I_0 \exp(-\mu_{12}x-\mu_{22}x)$


Detector
Rotation axis
Object
X-ray source





a — 2D image

b — 3D CT slice

3D imaging

Structural

X-ray
Computed
Tomography

X-ray CT

Functional

Single
Photon
Emission
Computed
Tomography

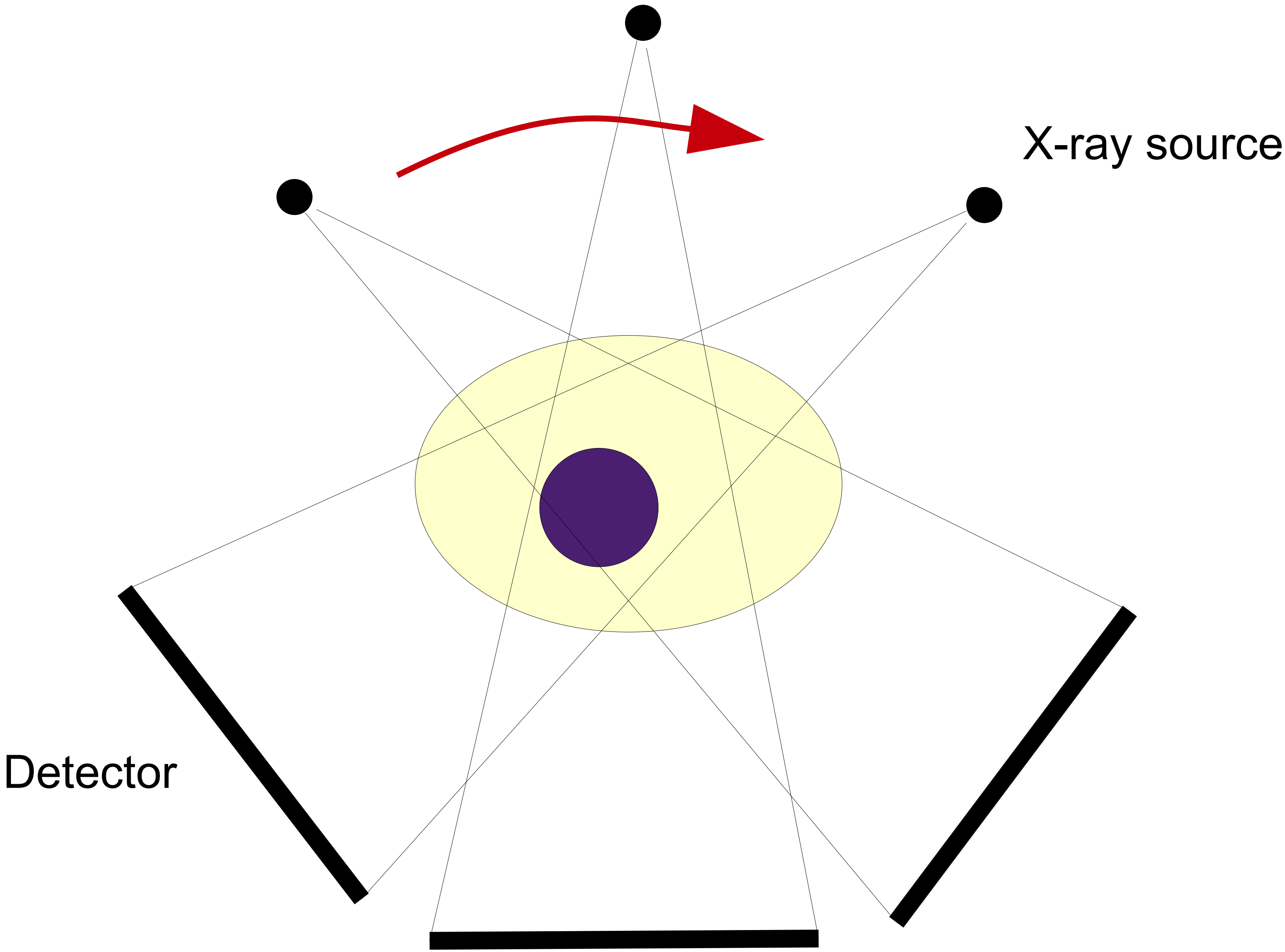
SPECT

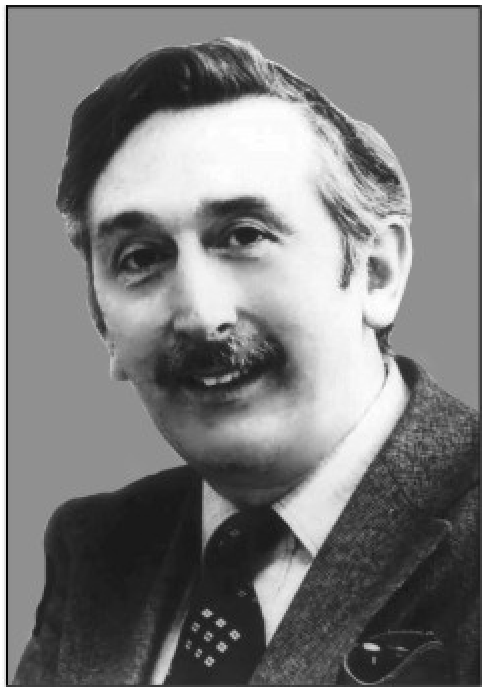
Positron
Emission
Tomography

PET

Xray-CT: transmission tomography

Reconstructing slice from projections





A. Cormack and G.Hounsfield

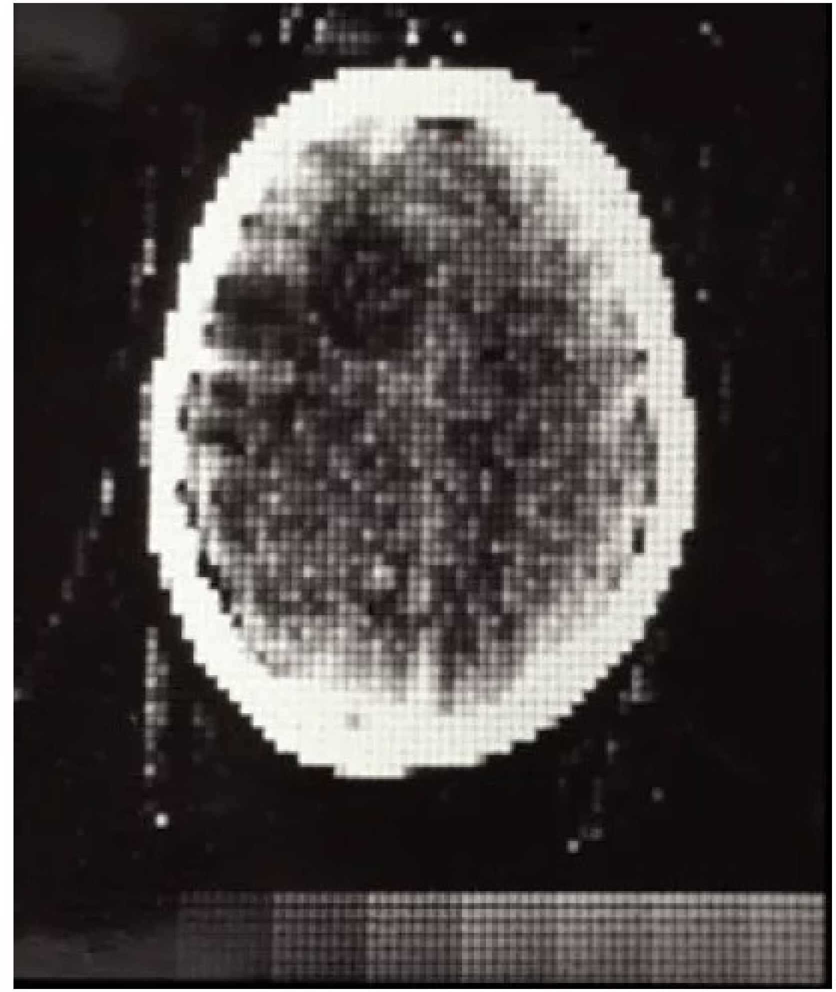
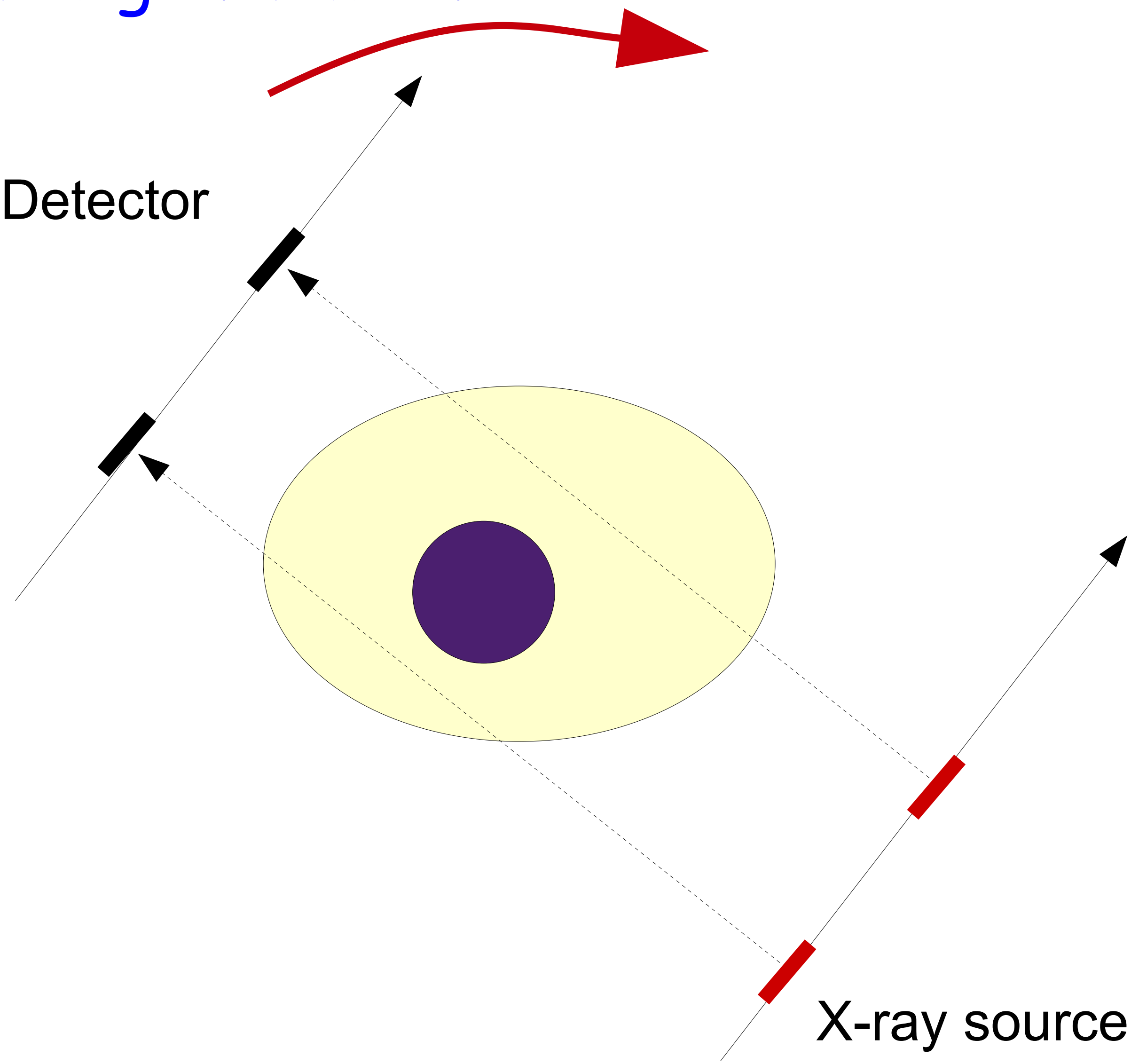


Figure 2. The first clinical scan: Atkinson Morley's Hospital, October 1971.

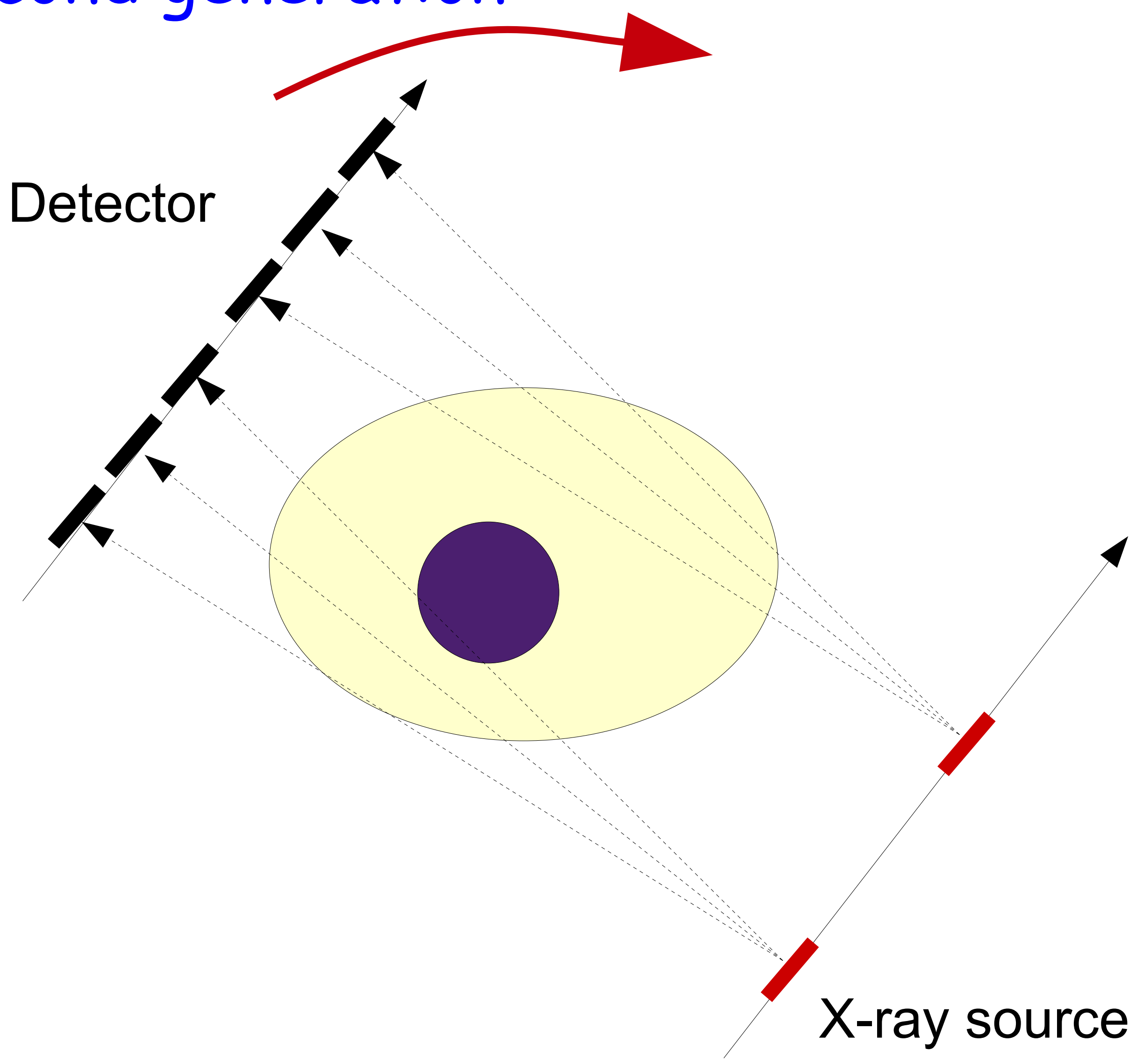
First generation

- Pencil beam
- Single detector
- Translate-rotate acquisition
- Very slow (minutes per slice)



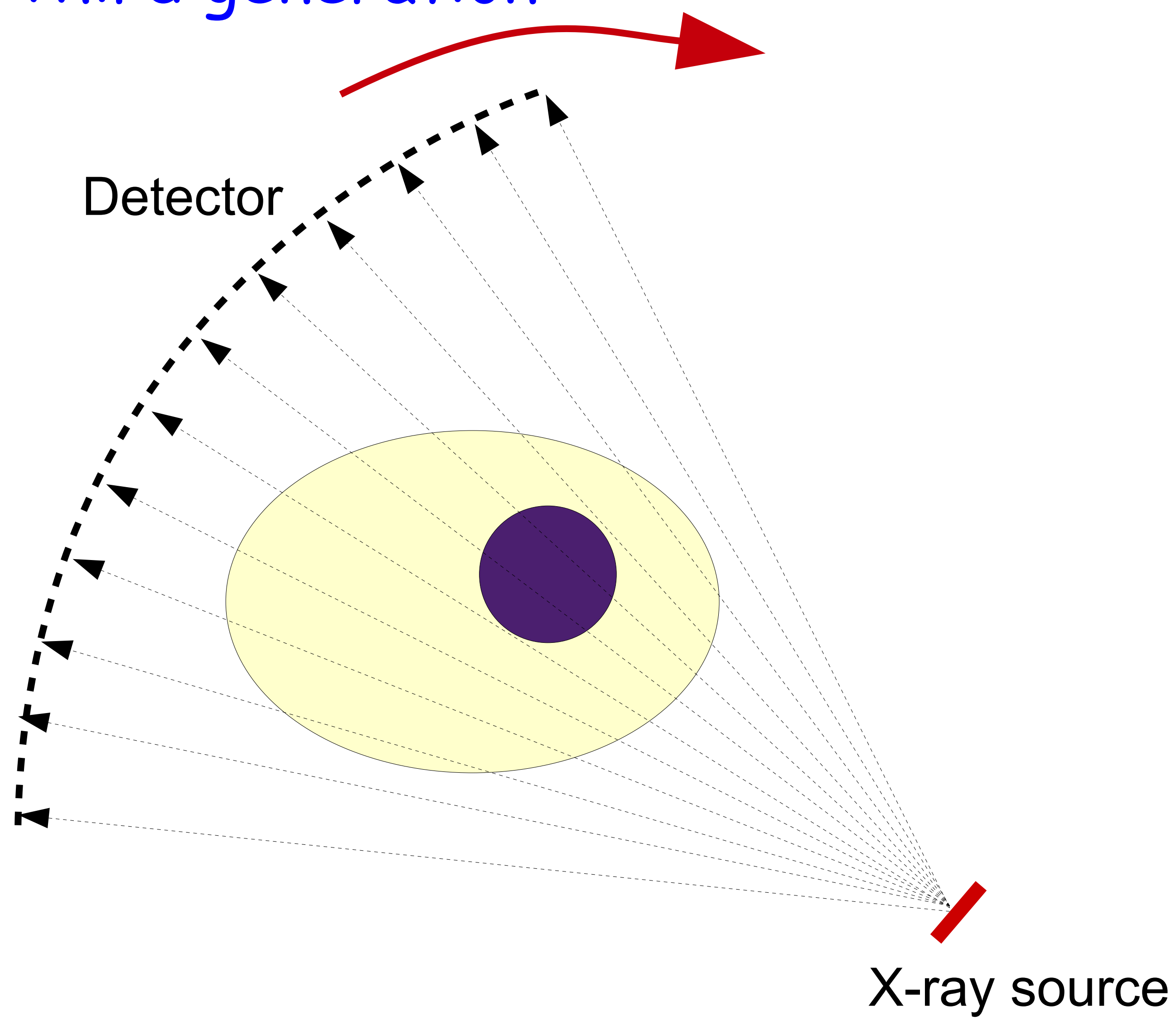
Second generation

- Narrow fan beam
- Multiple detectors
- Translate-rotate acquisition
- Slow (~20 s per slice)



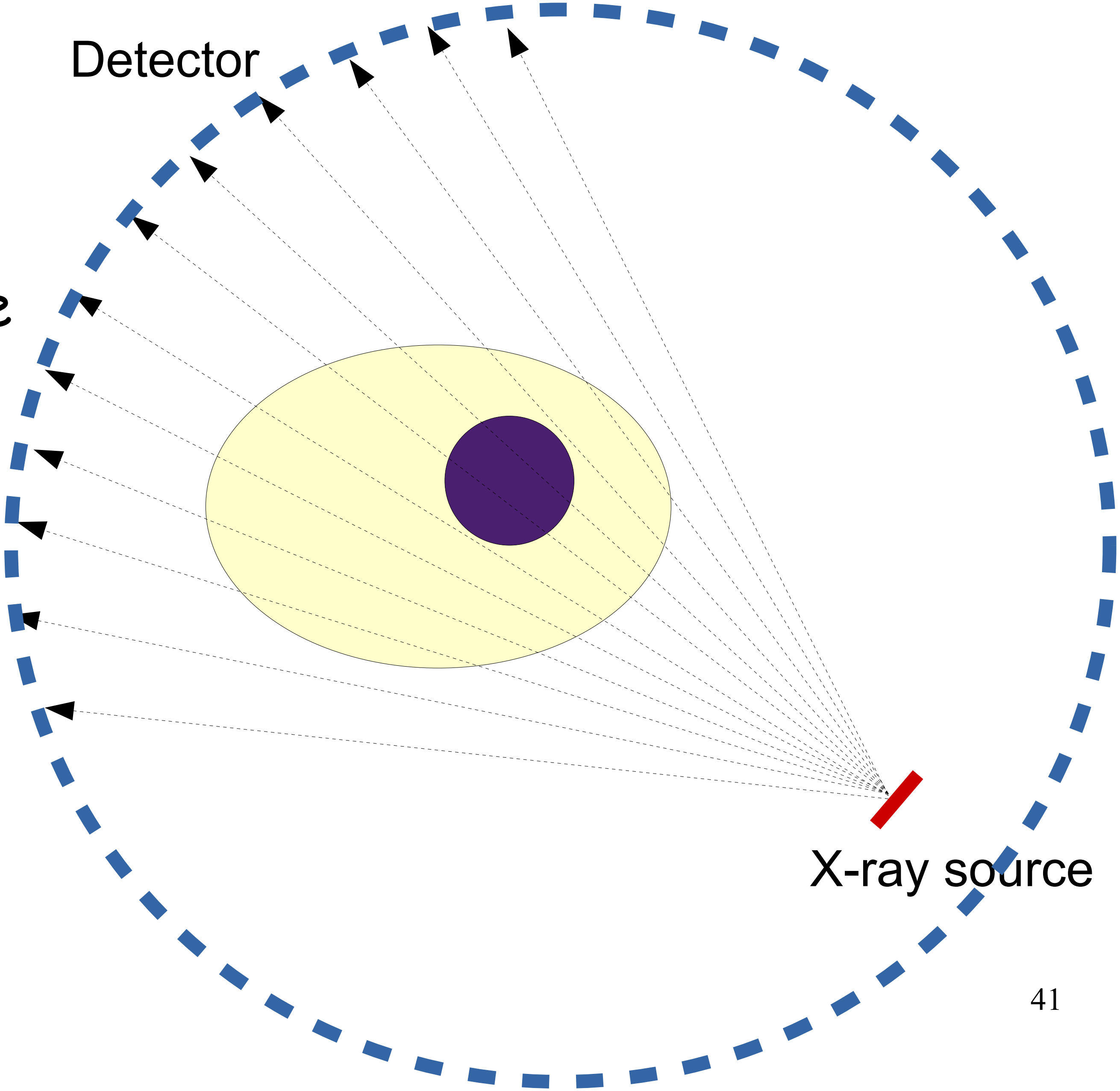
Third generation

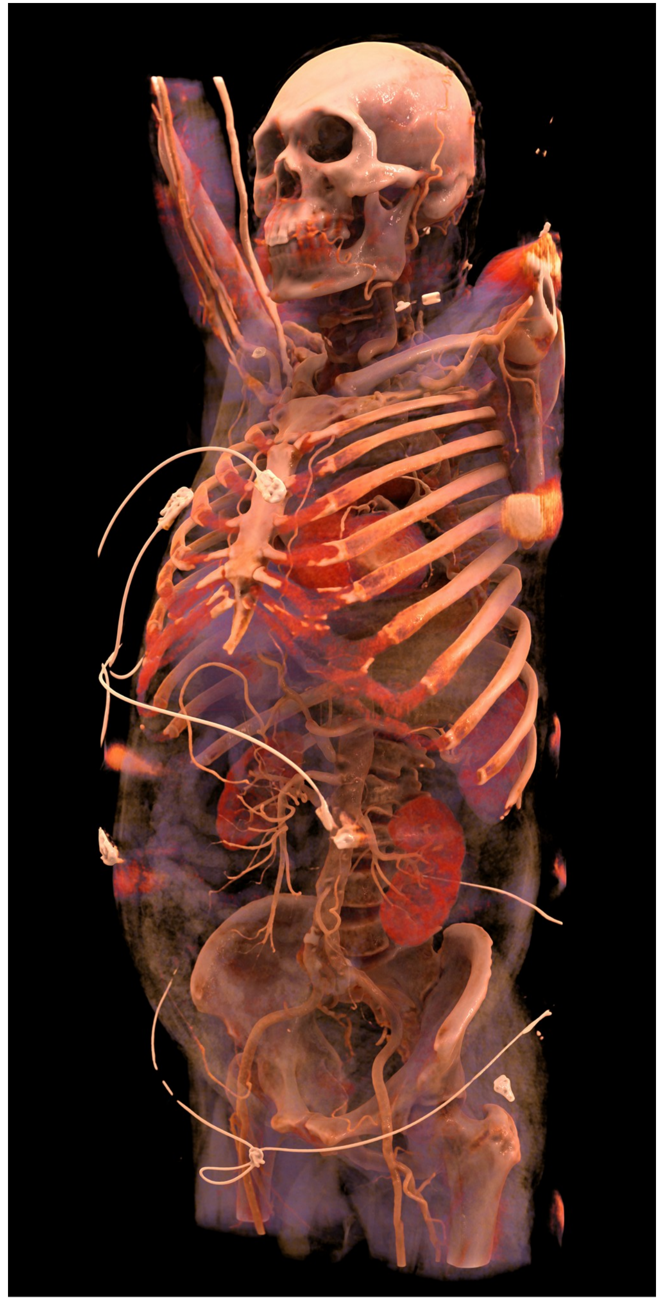
- Fan or cone beam
- Multiple detectors (up to 1000)
- Multiple detector rows (up to 64, multislice CT)
- No translation
- Fast (~0.5 s per slice)



Fourth generation

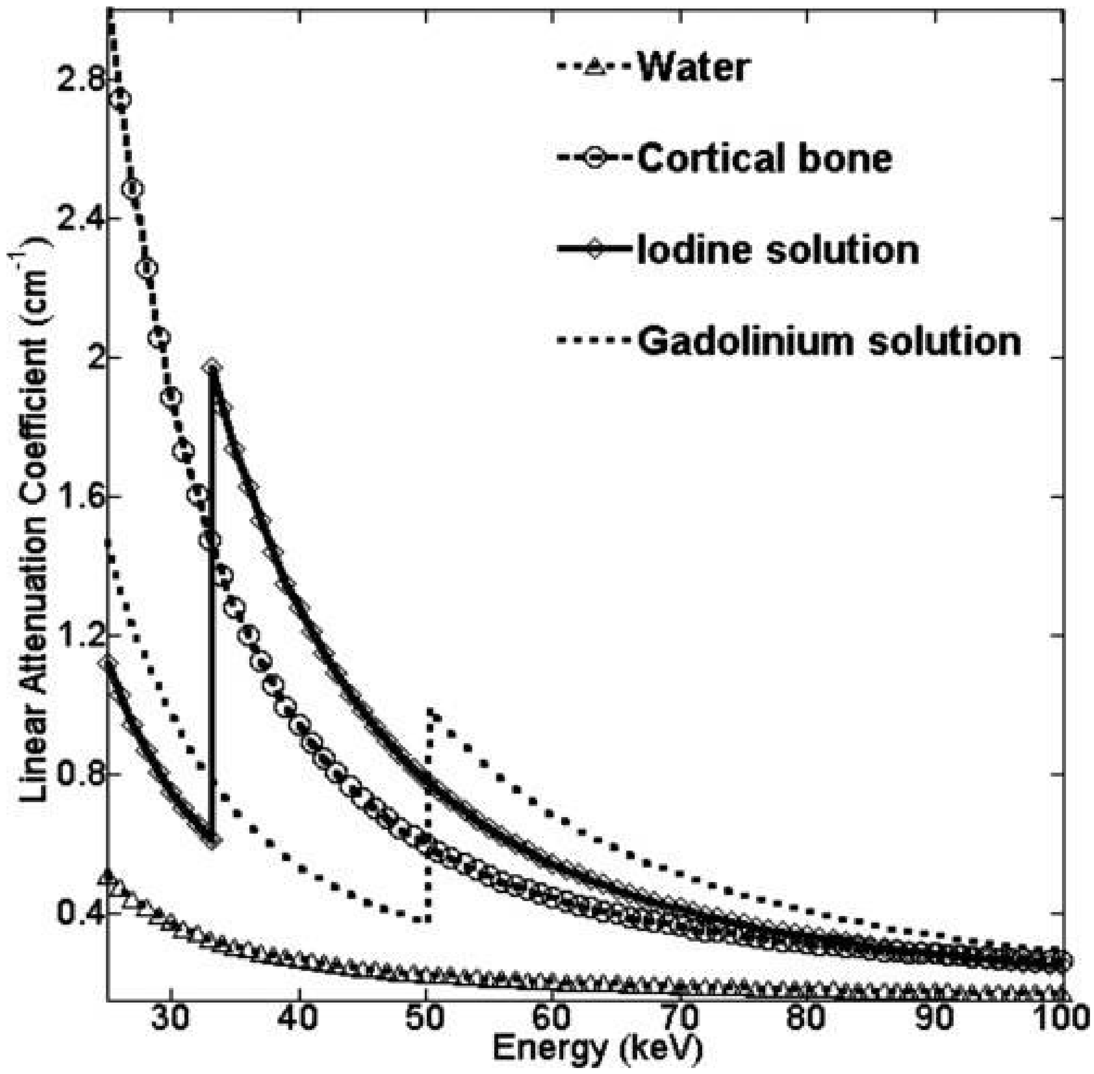
- Multiple detectors
- Only X-ray source rotates
- Fast (~0.5 s per slice)

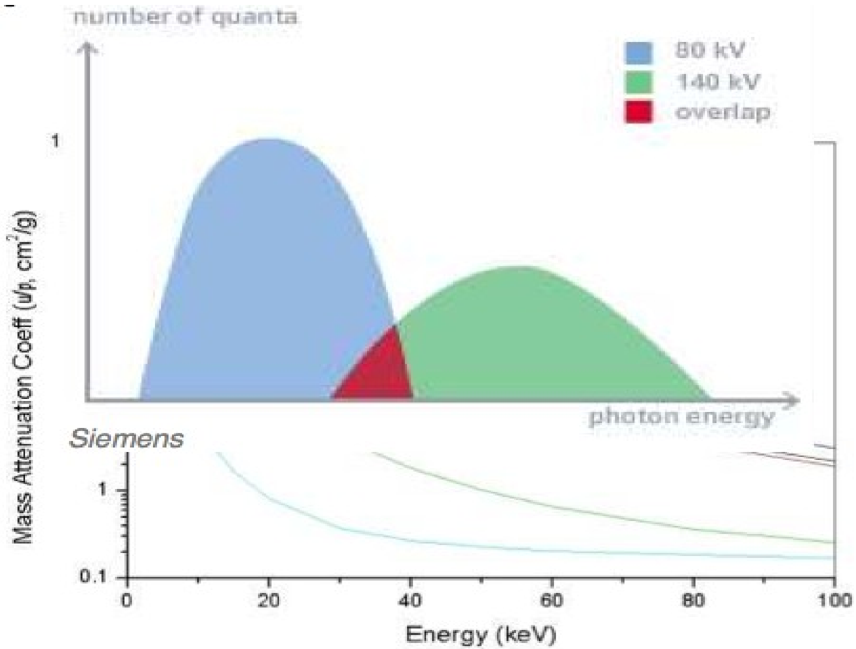




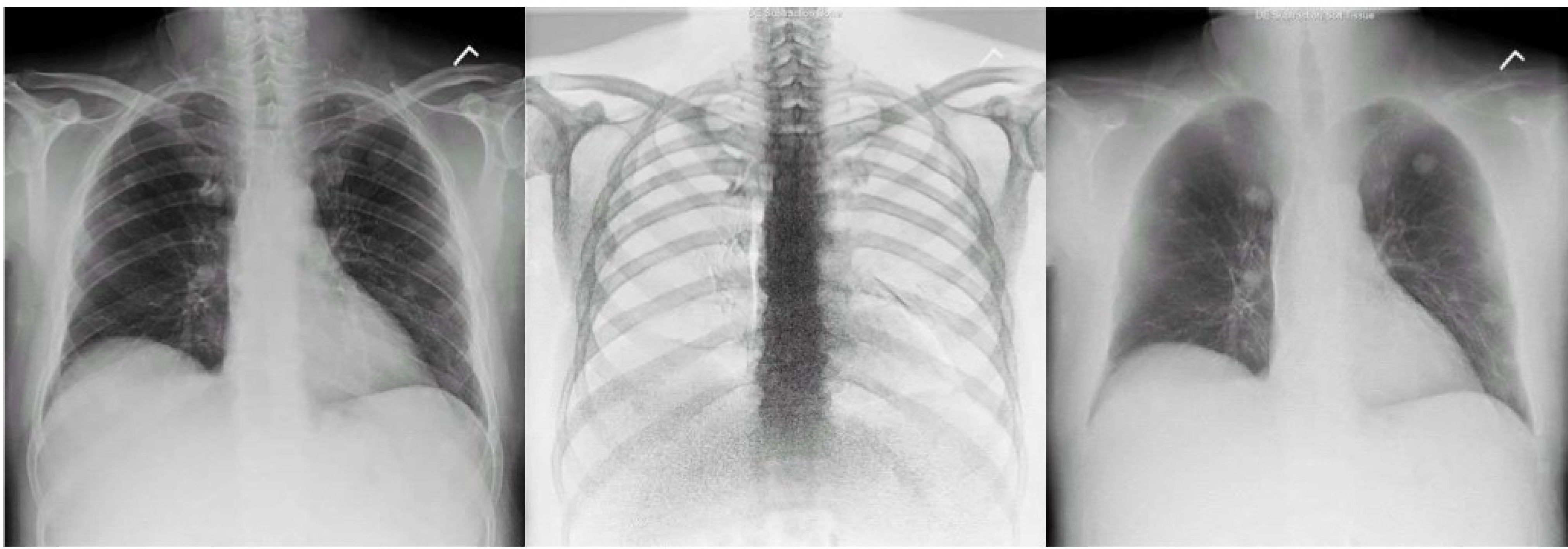
Courtesy of Siemens Healthineers

What if we look at the photon energy?



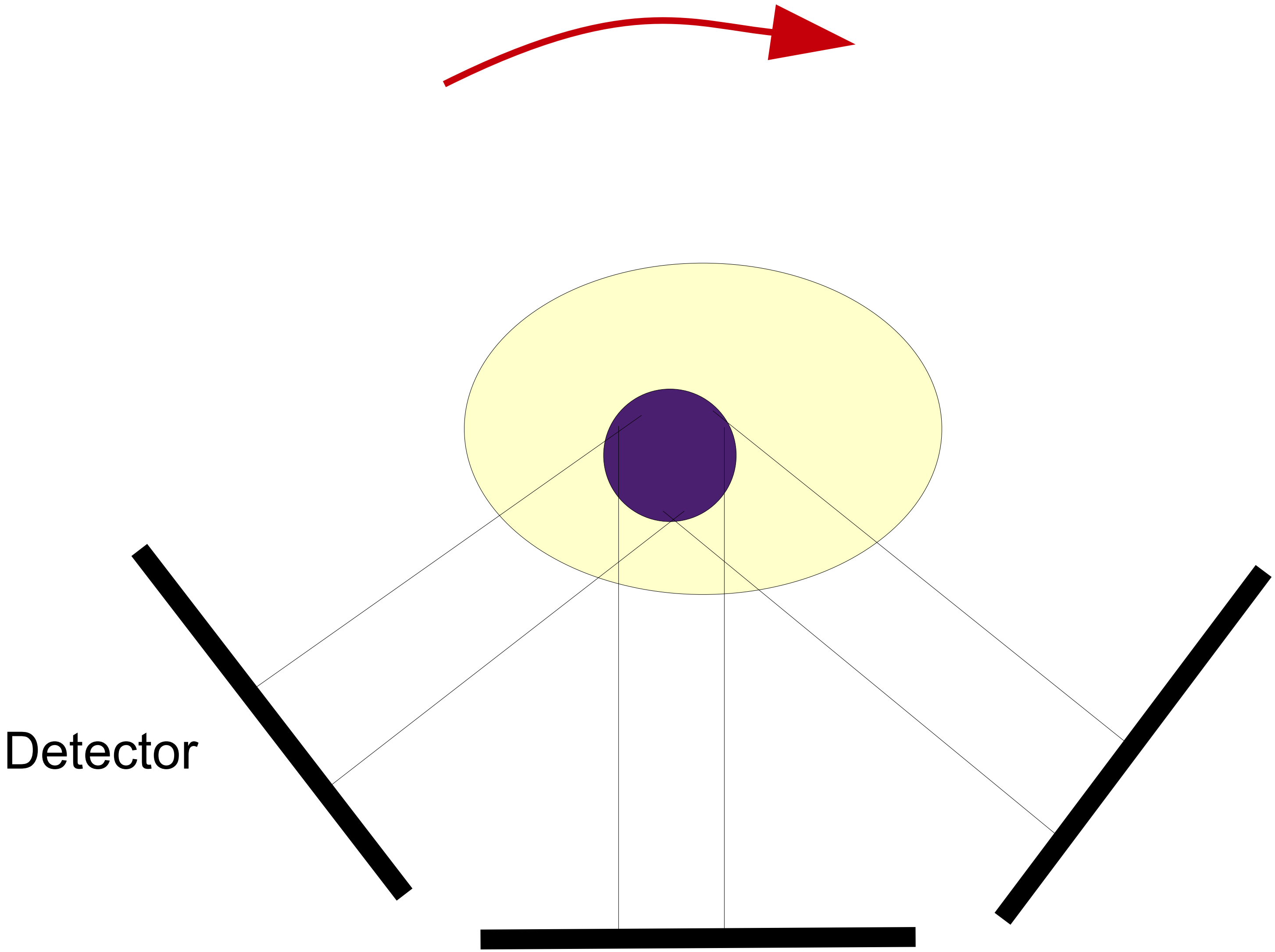


An example of dual energy CT (Siemens SOMATOM Definition)

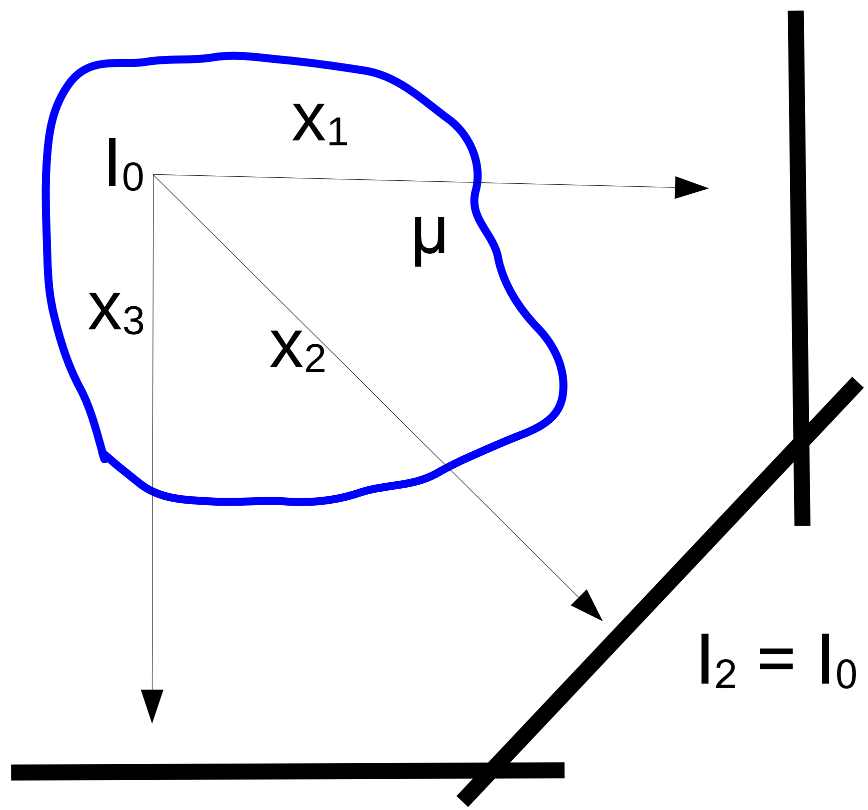


SPECT: emission tomography

Reconstructing source position and activity from projections



$I = I_0 \exp(-\mu x)$



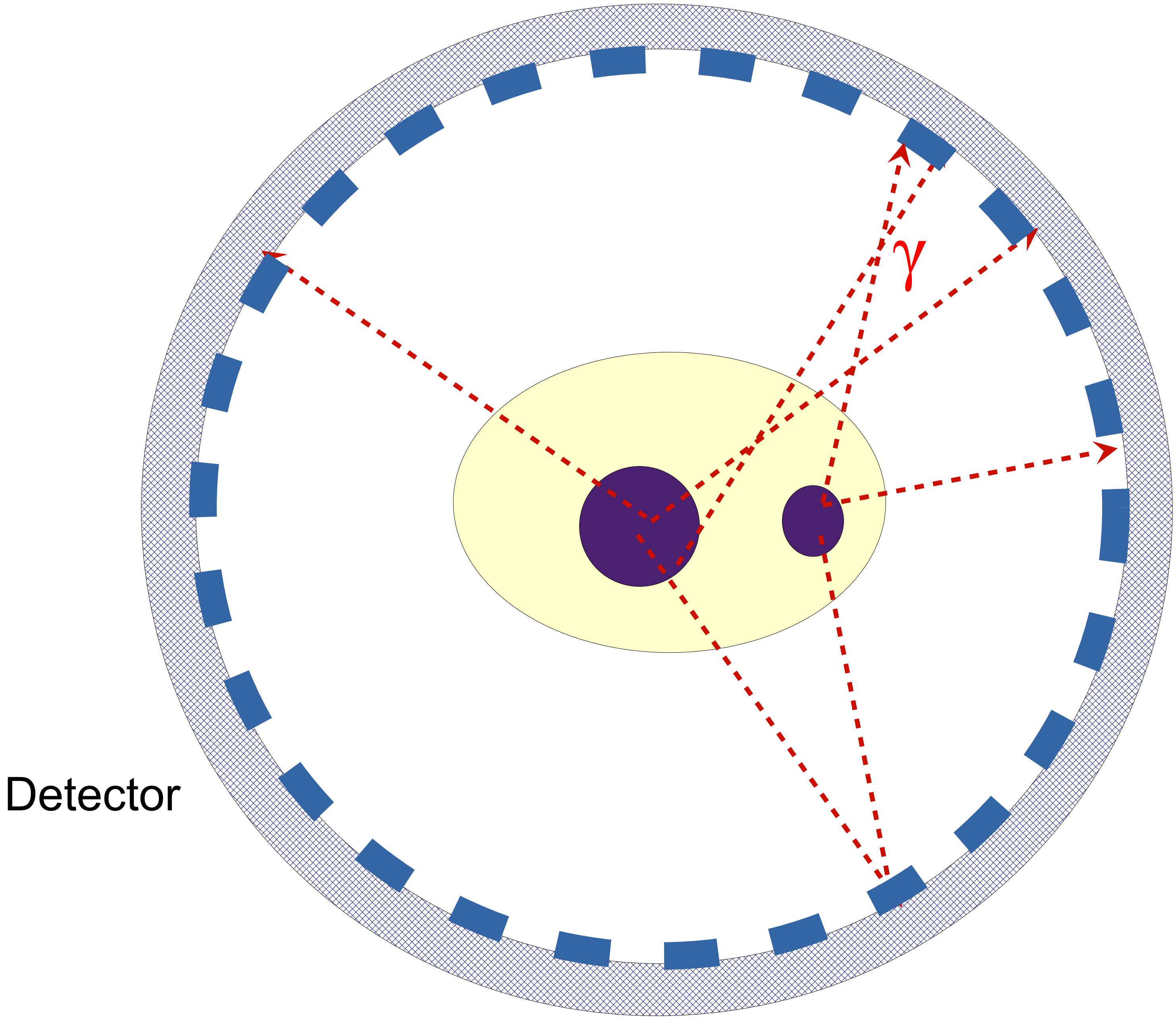
$I_1 = I_0 \exp(-\mu x_1)$

$I_2 = I_0 \exp(-\mu x_2)$

$I_3 = I_0 \exp(-\mu x_3)$

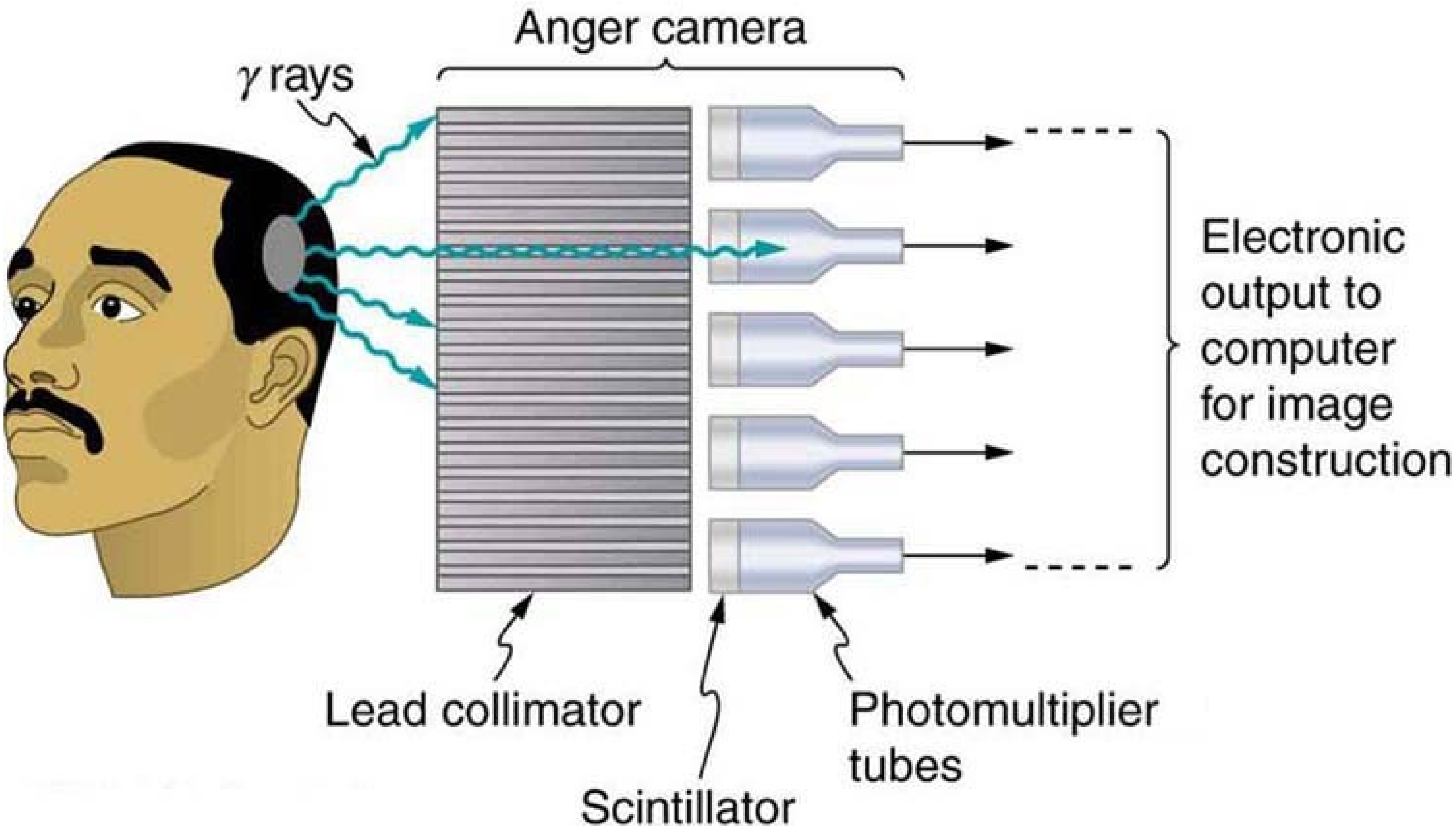
It is necessary to know the photon direction

SPECT: Single Photon Emission Computed Tomography

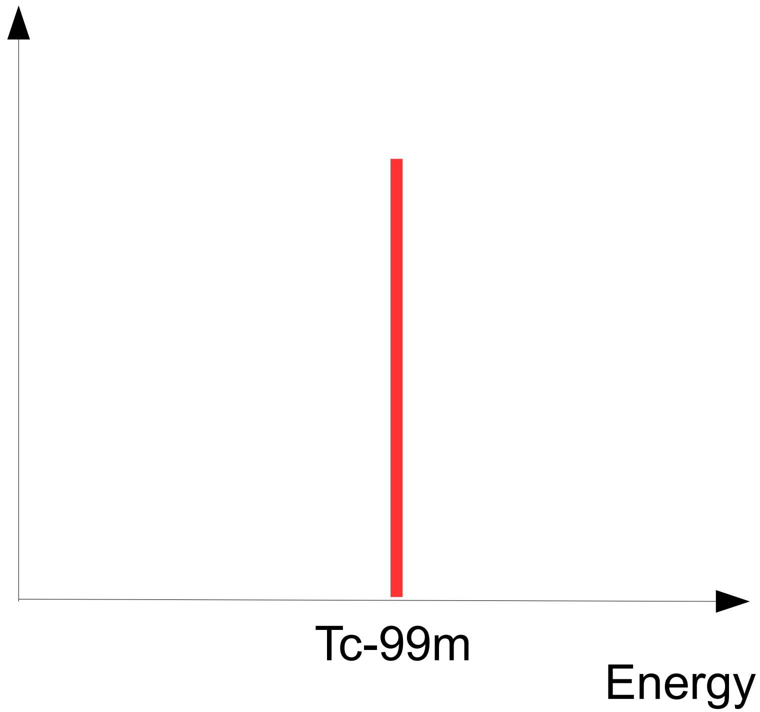


Gamma camera

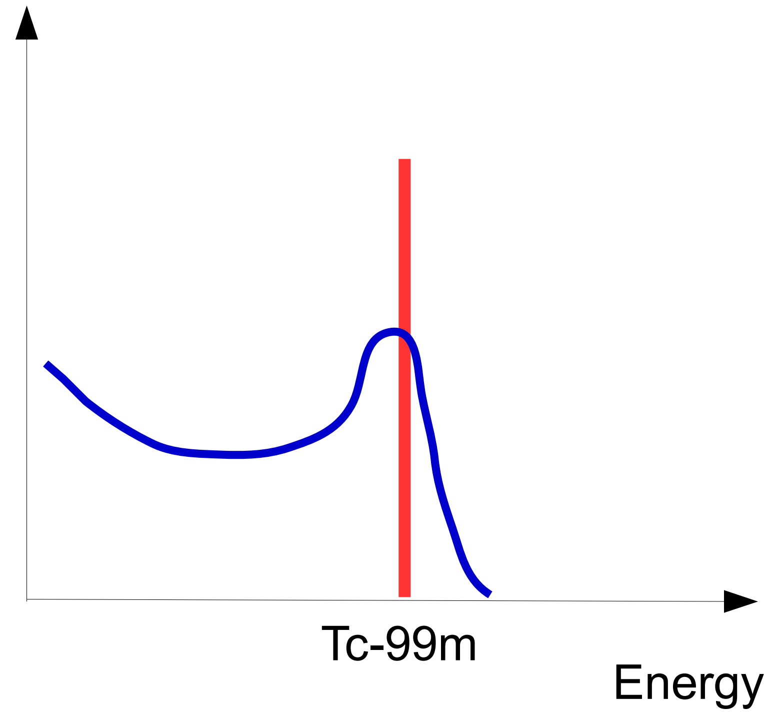
H. Anger, 1958



A problem



Ideal gamma spectrum



Real gamma spectrum:
Compton scattering

Background is higher, sensitivity is lower

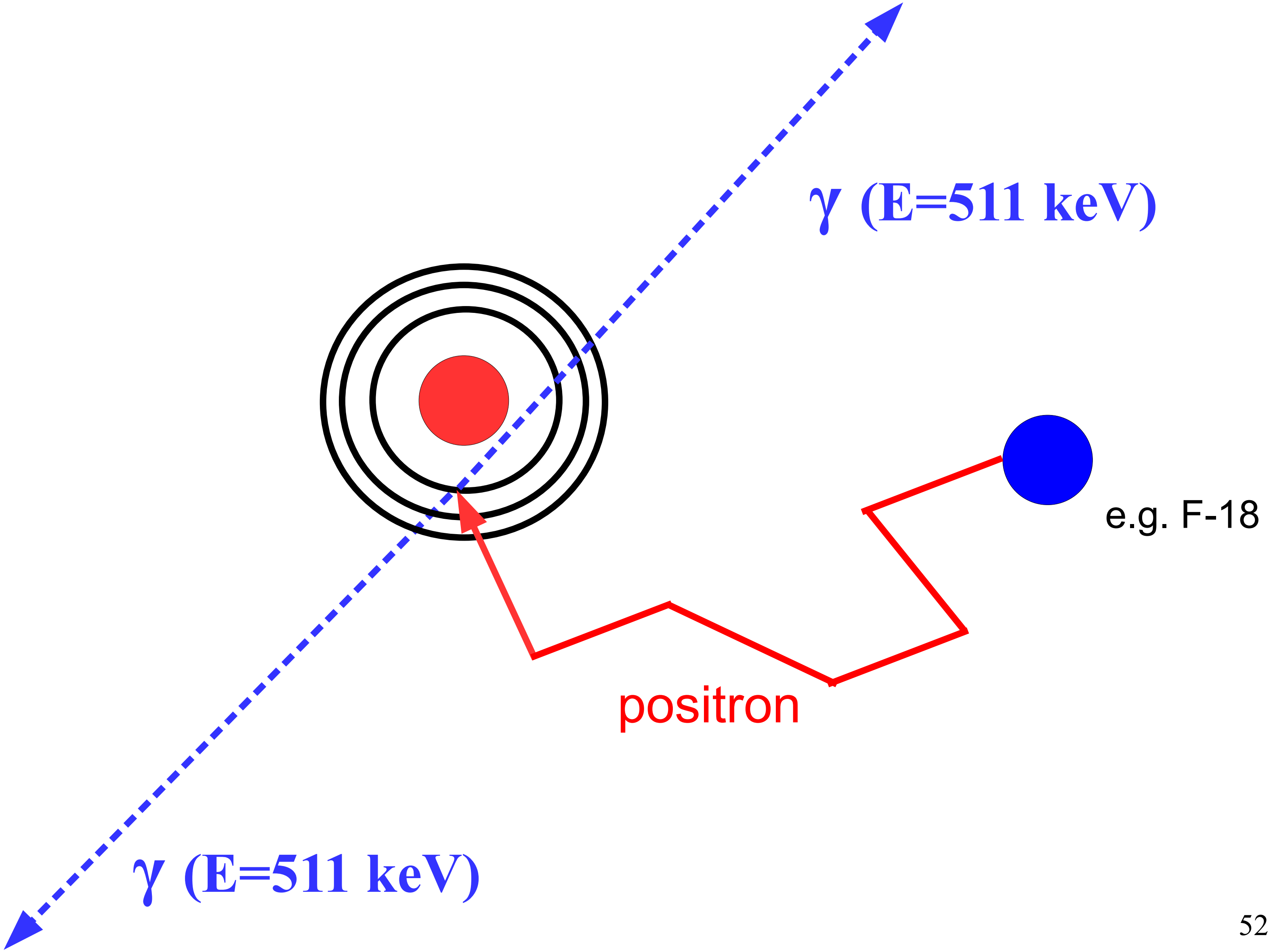
Energy resolution helps tp clean up the picture

SPECT detectors

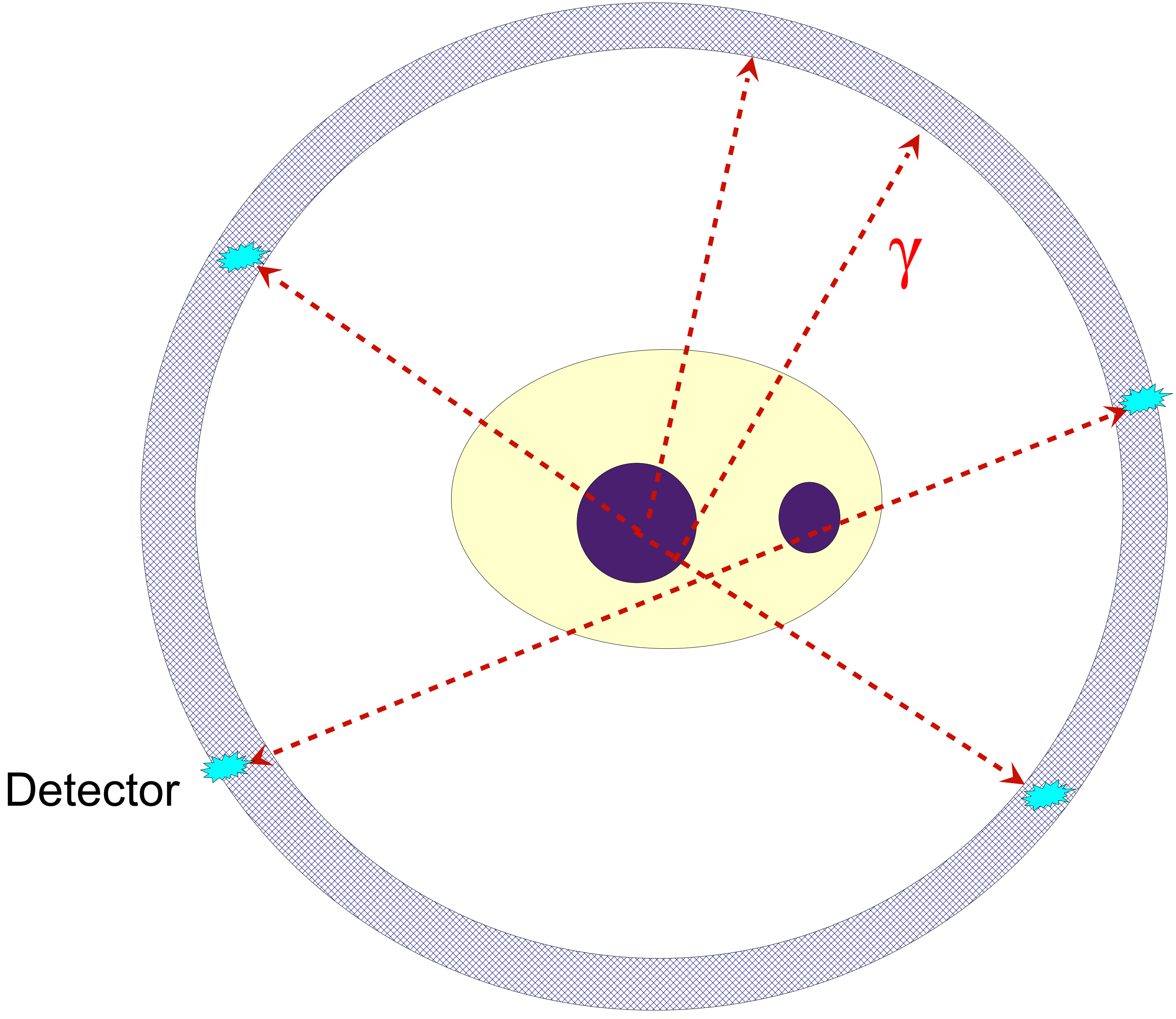
- Scintillator detectors (NaI(Tl) is the most widespread so far – optimal for Tc-99m, large crystals)
- Good energy resolution in the range of 30-250 keV is necessary (suppress background, dual-isotope studies)
- Semiconductor detectors are promising due to their energy and spatial resolution
  - detection efficiency is a challenge
  - spatial resolution (~0.5 – 1 mm) is limited by collimators

more details in 2011 Phys. Med. Biol. 56 R145

PET: Positron Emission Tomography

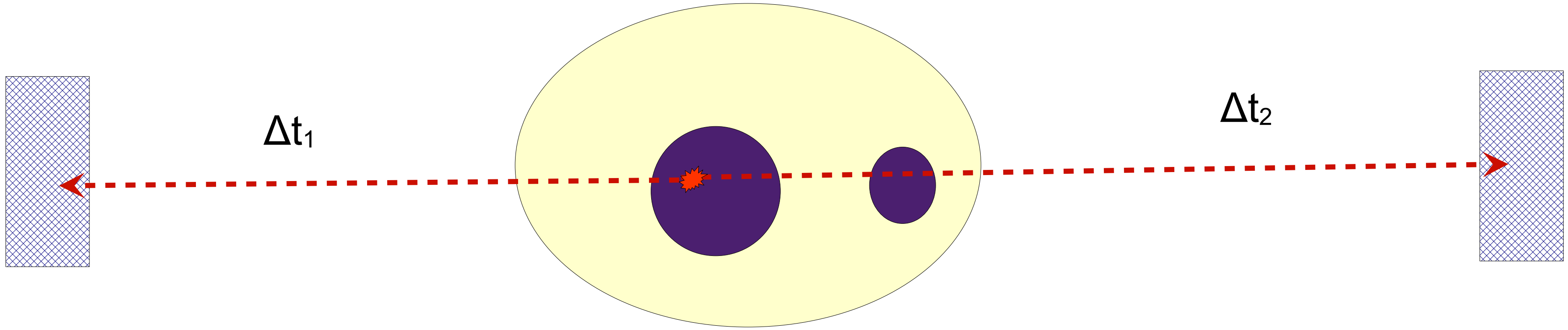


PET



PET detector features

- No collimator is needed
- Optimized to detect 511 keV gamma (higher energy) – heavy scintillators (BGO, LSO) => spatial resolution
- Energy resolution ~10%
- Time resolution <0.5 ns – TOF-PET:



TOF-PET

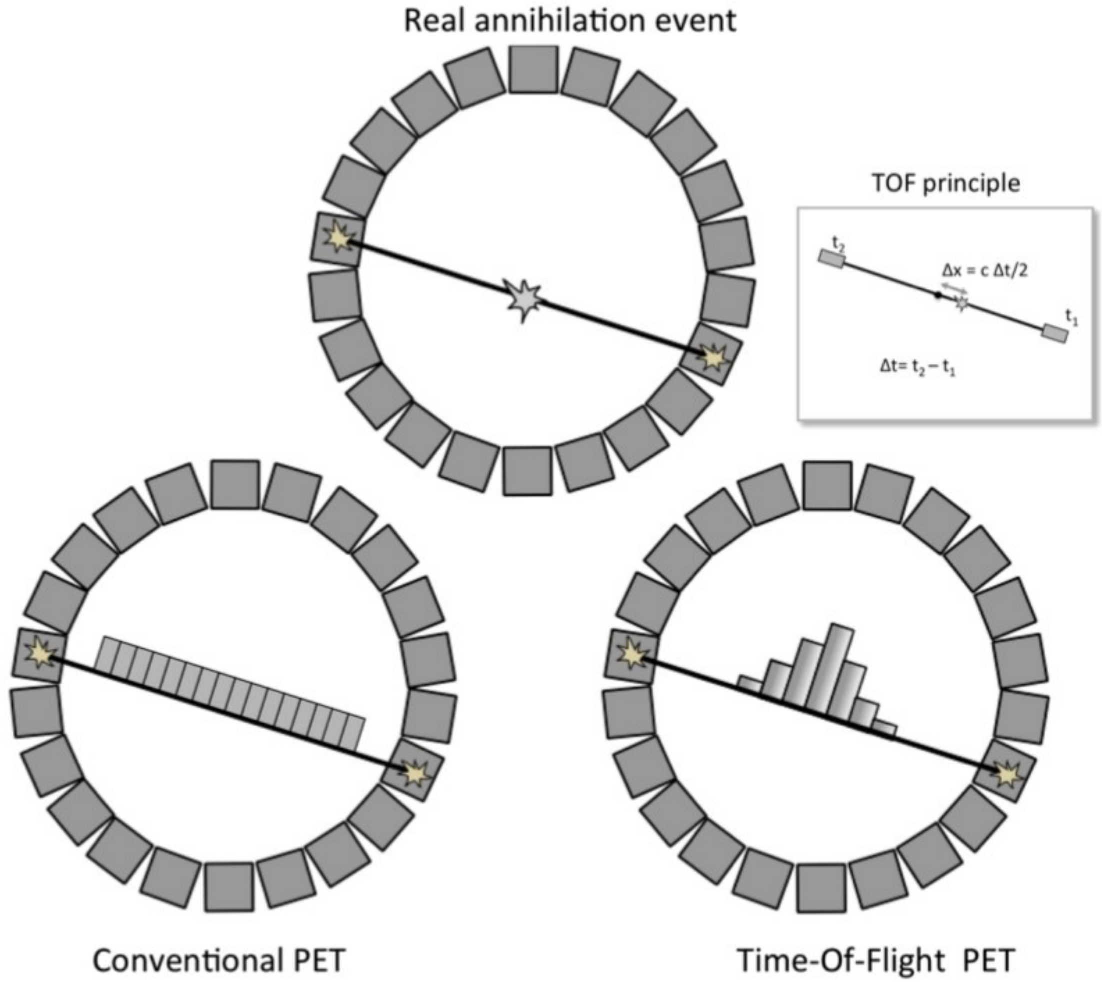


Fig. 1 Compared to conventional PET, the estimated time-of-flight difference (Δt) between the arrival times of photons on both detectors in TOF-PET allows localization (with a certain probability) of the point of annihilation on the line of response. In TOF-PET, the distance to the origin of scanner (Δx) is proportional to the TOF difference via the relation: Δt: $\Delta x = \frac{c\Delta t}{2}$, where c is the speed of light. t_1 is the arrival time on the first detector, and t_2 is the arrival time on the second detector

Vandenberghe et al. EJNMMI Physics(2016) 3:3

SPECT vs. PET

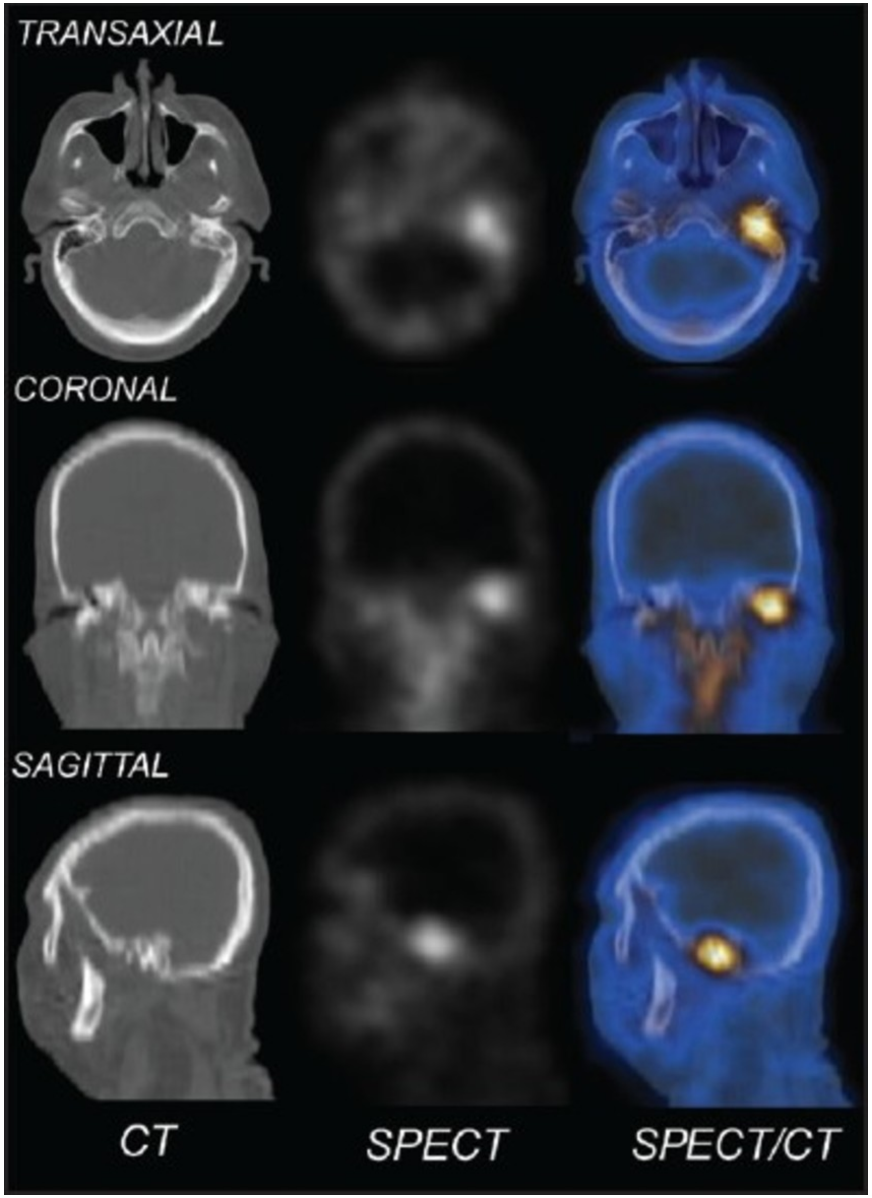
SPECT

- modest infrastructure requirements
- larger number of available radiotracers
- less sensitive than PET
- spatial resolution is rather low (few mm)

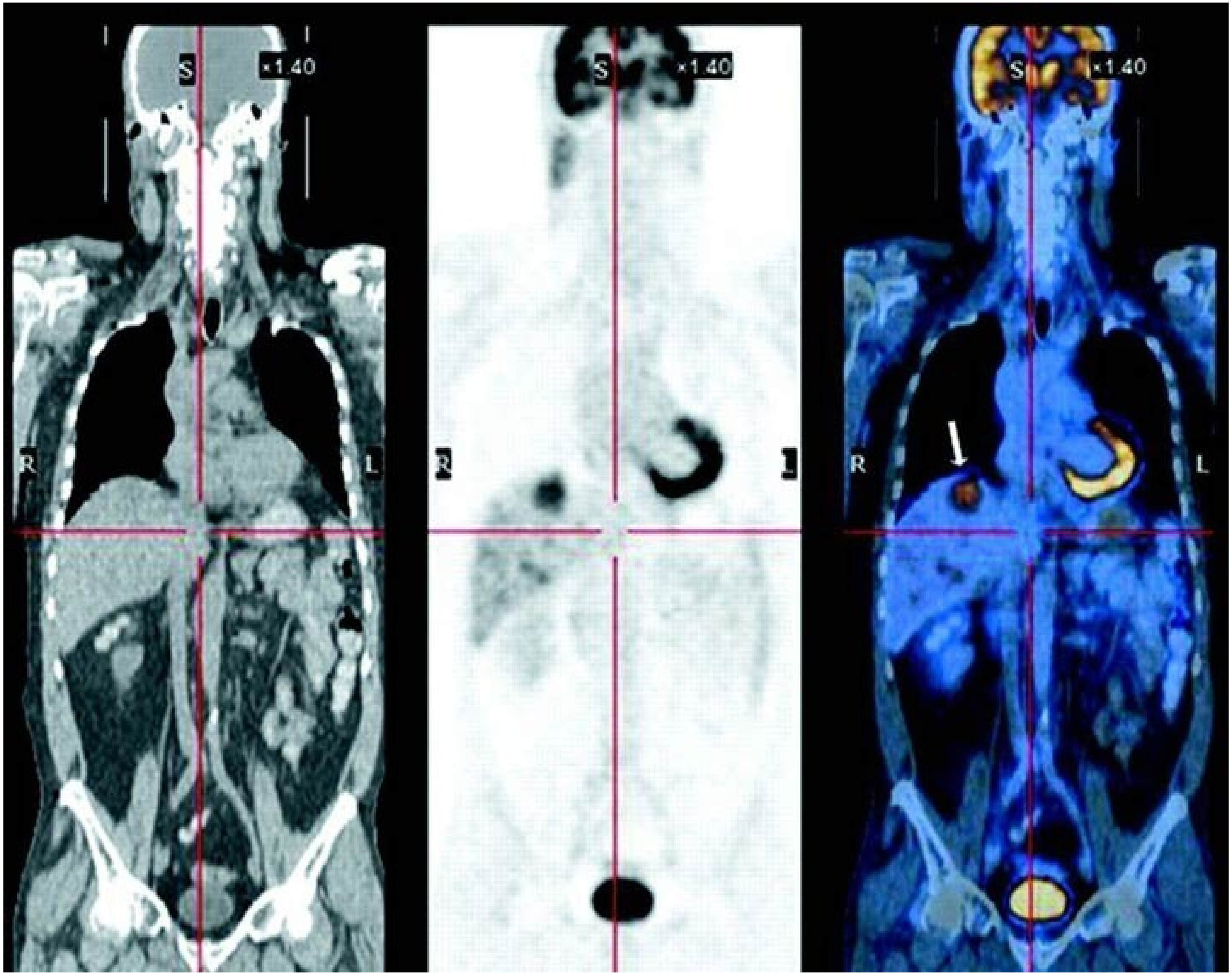
PET

- the most sensitive functional imaging method so far
- expensive (cyclotron)
- spatial resolution is comparable to SPECT

Combined structural and functional imaging: SPECT/CT



Combined structural and functional imaging: PET/CT



Future prospects & challenges

- **CT**
  - Higher spatial resolution, single photon counting, spectral CT
- **SPECT(/CT)**
  - whole-body SPECT, higher spatial resolution, new radionuclides
- **PET(/CT)**
  - Higher time resolution (up to 10 ps), long axial FOV (up to 1-2 m)
- **AI for all applications**
  - Detector calibration, mage preprocessing, reconstruction, image postprocessing, low statistics signals...
  - Generalization problem and the bias control
- **Computing for all applications**
  - amount of data grows quadratically with the spatial resolution ...
- **Multiple modality (PET/CT, SPECT/CT, PET/SPECT/CT**
- **Cost and economy**
  -

Summary

- Xray-CT, SPECT and PET are very powerful methods for medical imaging nowadays
- Scintillator detector remains a main working horse for photon detection. Prospects of semiconductors are still rather vague
- Constant hardware innovations:
  - finer spatial resolution (= more details), larger FOV
  - finer energy resolution for SPECT and PET, spectral CT
  - precise timing for TOF-PET
  - higher sensitivity (=lower dose, early diagnosis) — lower noise
  - speed for X-ray CT (very important for cardiac CT)
- Significant challenges arise in data processing: AI and high performance computing