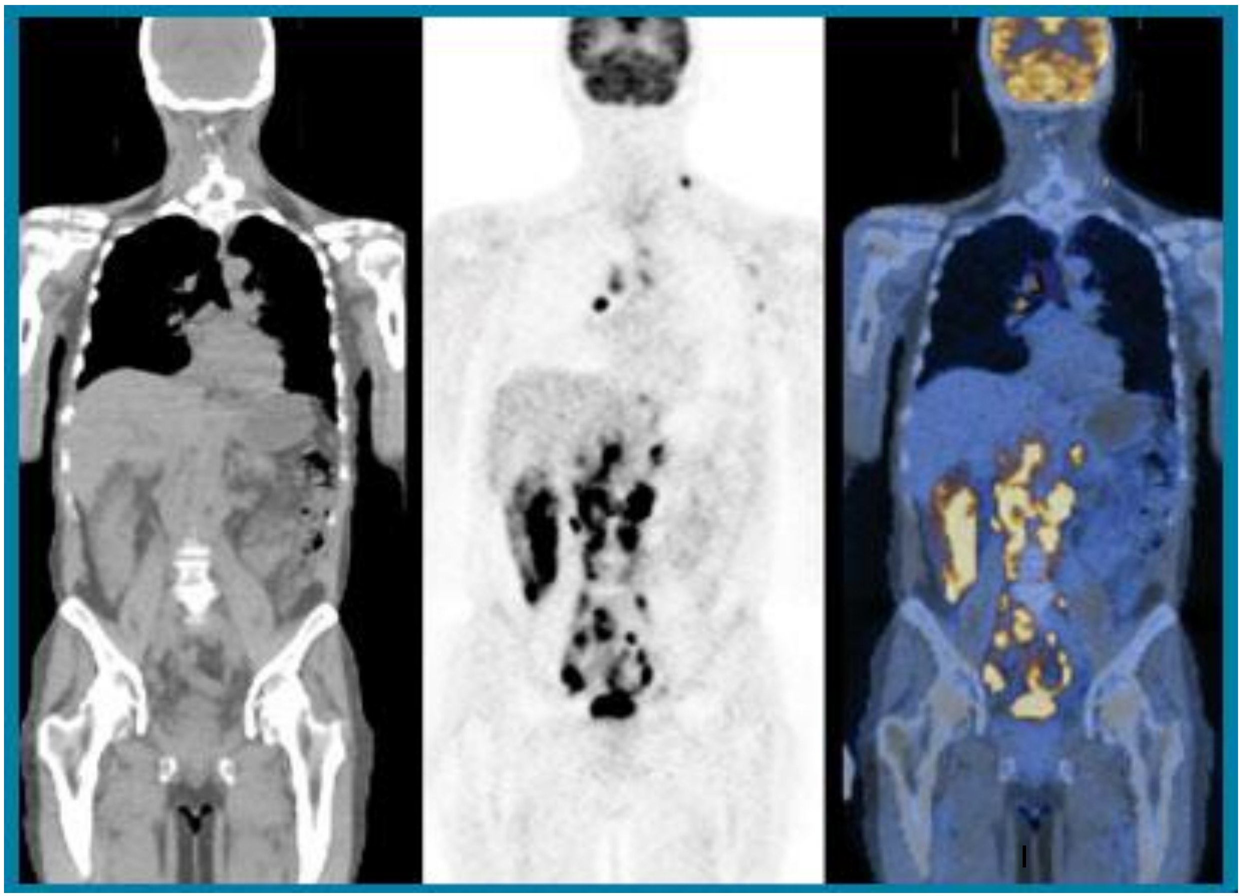
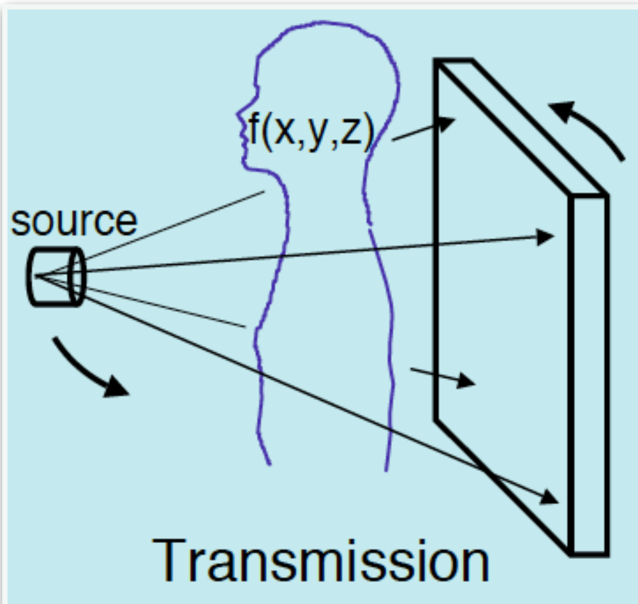
Development of pixelated detectors in medicine

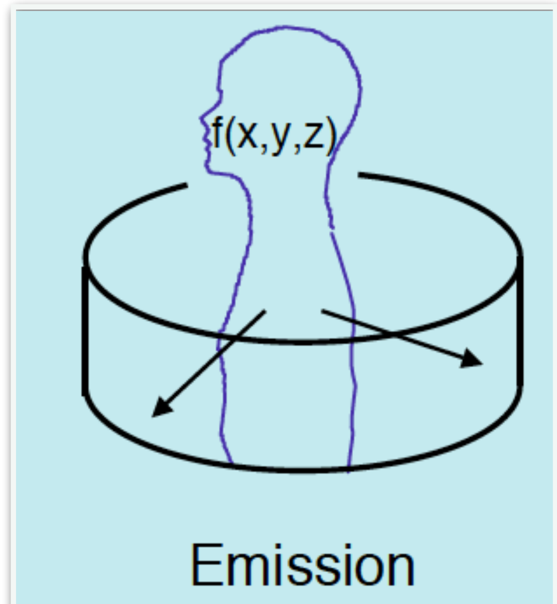


Prof. Dr. Erika Garutti

Universität Hamburg
DER FORSCHUNG | DER LEHRE | DER BILDUNG

Look inside the human body

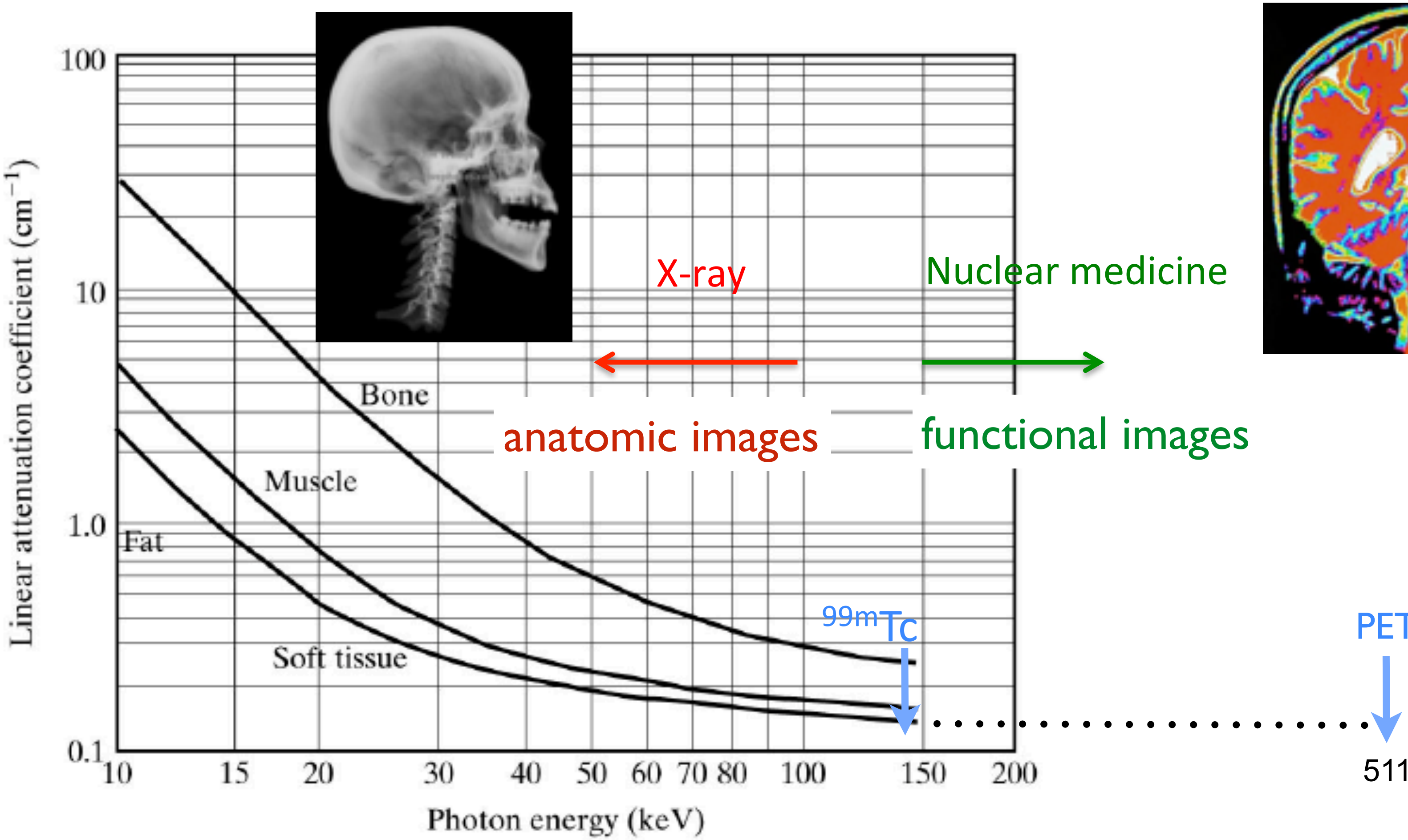




- absorption of particles proportional to material density
- anatomical structure
- X-ray photons: 10-100 keV

- rate of emissions depends on concentration of particles
- concentration related to the functional activity of cells
- Gamma photons: 100-511 keV

Photon attenuation in tissues



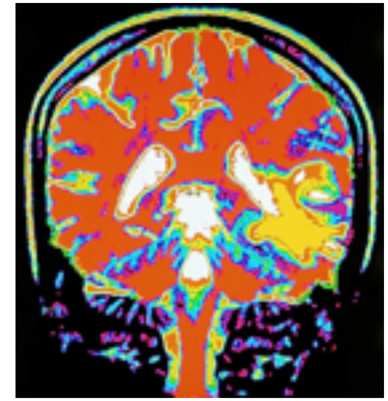
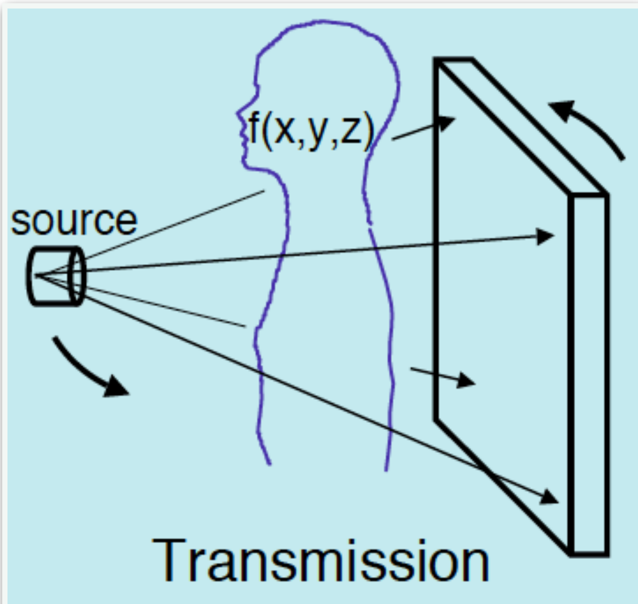
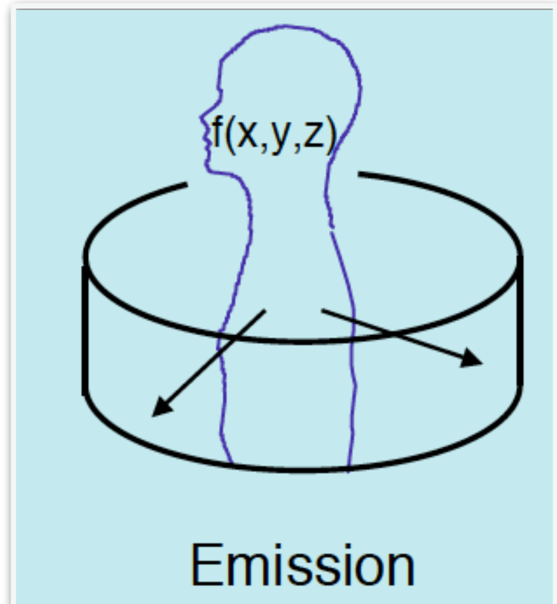


Figure 4.8

Medical Imaging Signals and Systems, by Jerry L. Prince and Jonathan Links.

Detector requirements





- X-ray photons: 10-100 keV
- broad energy spectrum
- high rate (~10^9 photon/mm^2/s)
- large number of pixels (>100k)

- measure (E integrated) photon flux
  … single photon counting & E wished

- Gamma photons: 100-511 keV
- monochromatic energy + Compton
- low rate (~10^5 photon/mm^2/s)

- measure single photon energy & time of arrival

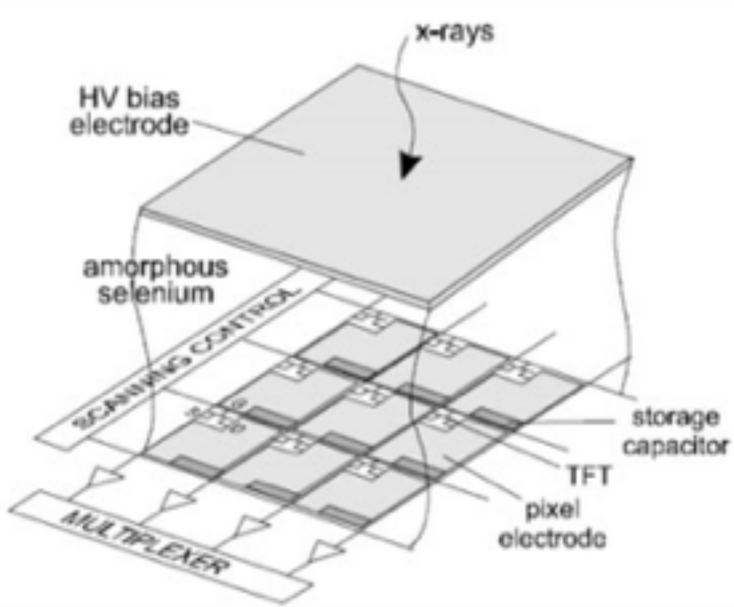
Detector approach

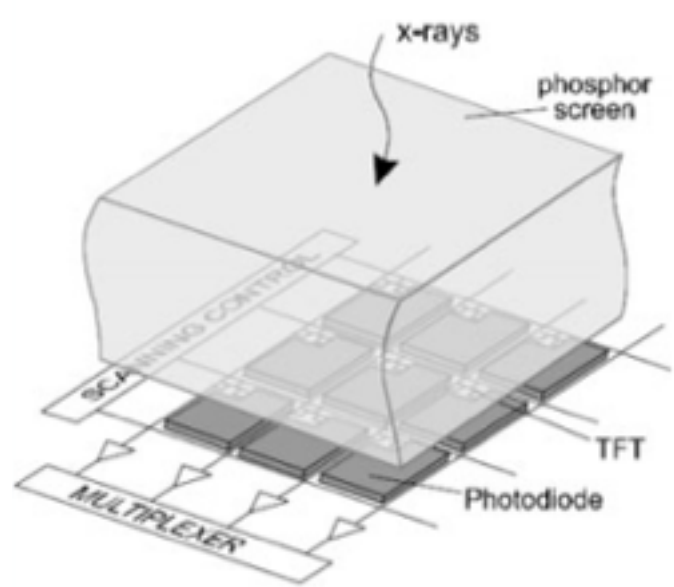
Direct detection:

conversion of photon energy deposited in a semiconductor to an electric charge

Indirect detection:

conversion of photon energy deposited in a scintillator to optical photons counted on a photodetector





TFT= Thin-Film Transistor

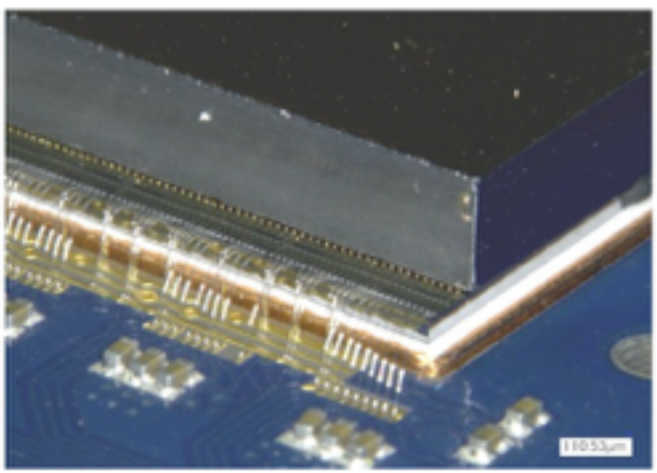
Reprinted with Rowlands. Flat panel detectors for medical X-ray: physics and technology. Available at: http://hepwww.rl.ac.uk/Vertex03/Talks/Row/Rowlands.pdf. Accessed December 20, 2010.[1]

Detector approach

Direct detection:

conversion of photon energy deposited in a semiconductor to an electric charge

Example: CdZnTe pixel detector (3mm thick / 250um pixels)



3mm CZT gold stud bonded to ASIC and wire bonded to CoB

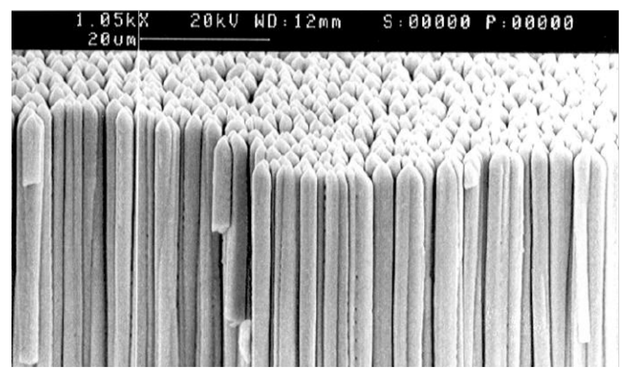
Paul Seller, RAL, PSD9

Indirect detection:

conversion of photon energy deposited in a scintillator to optical photons counted on a photodetector

Example: fluorescence screen

CsI:Tl needle crystals (500um thick / 5um diameter) + CCD



Detector approach

Direct detection:

conversion of photon energy deposited in a semiconductor to an electric charge

Indirect detection:

conversion of photon energy deposited in a scintillator to optical photons counted on a photodetector

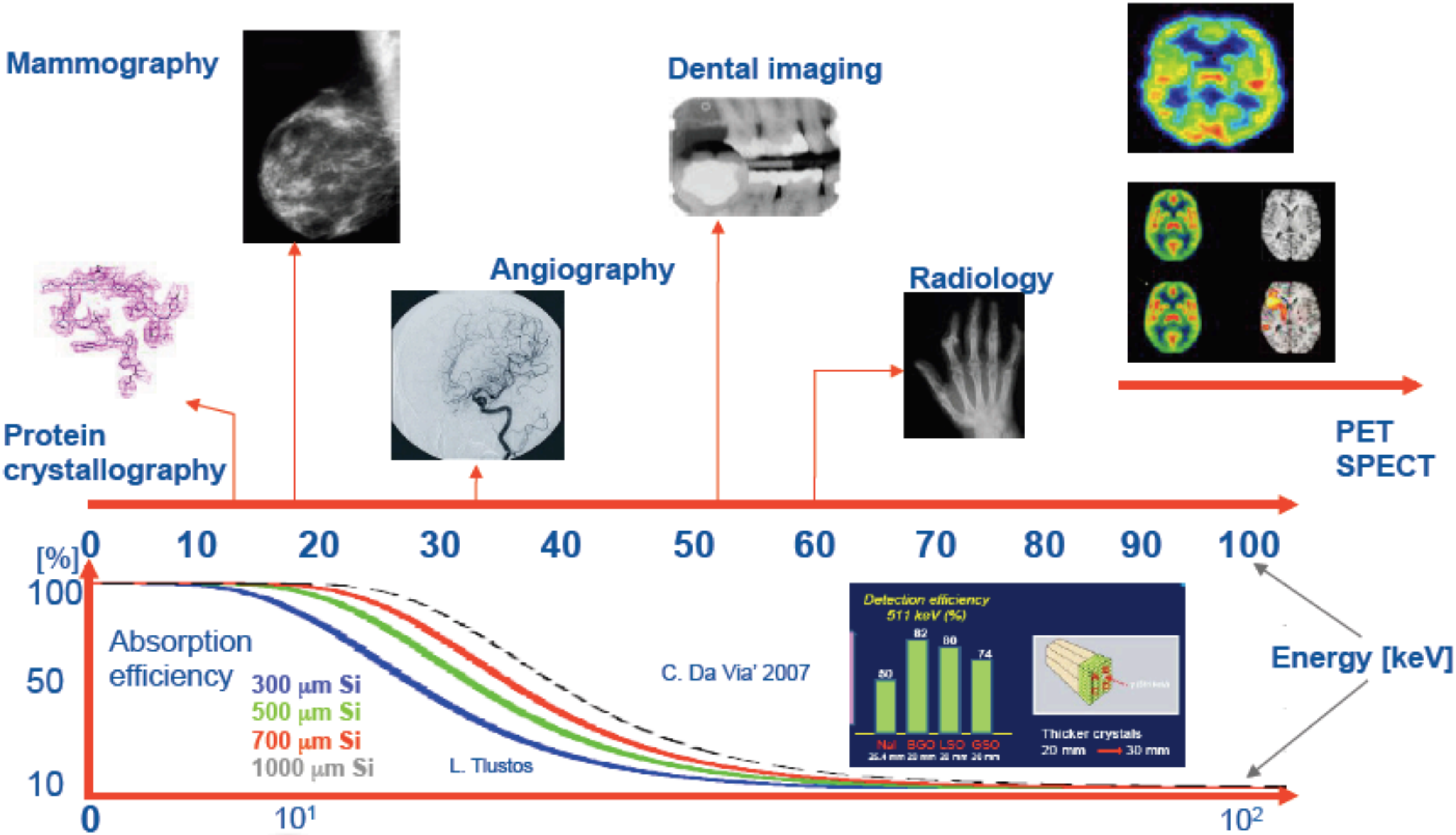
Direct vs Indirect Pro:

- factor ~10 lower conversion energy = better E resolution
- no lateral spread of signal (charge vs light) = better spatial resolution

Direct vs Indirect Cons:

- low density = low attenuation for high E photons = low sensitivity
- large quantity availability / cost

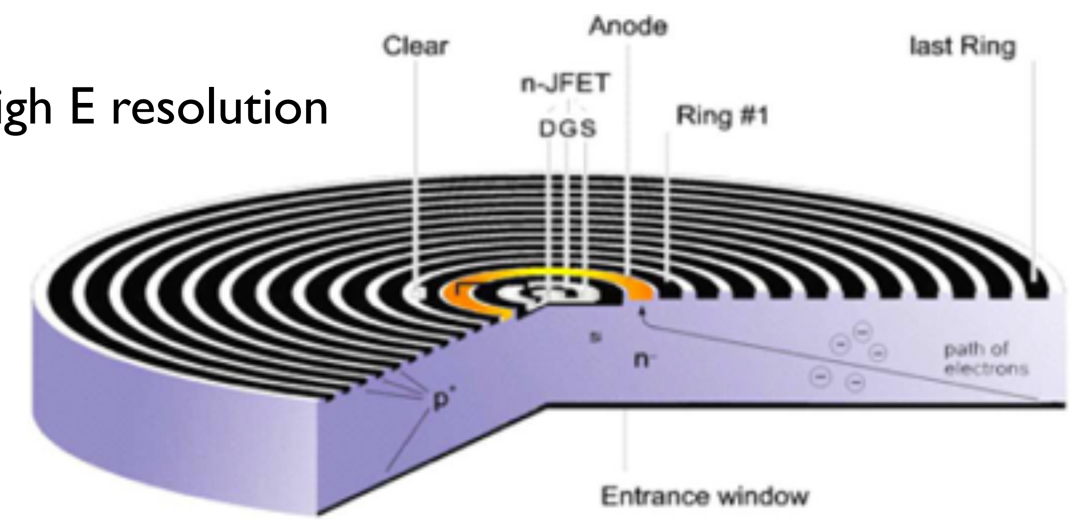
Possible materials: Silicon



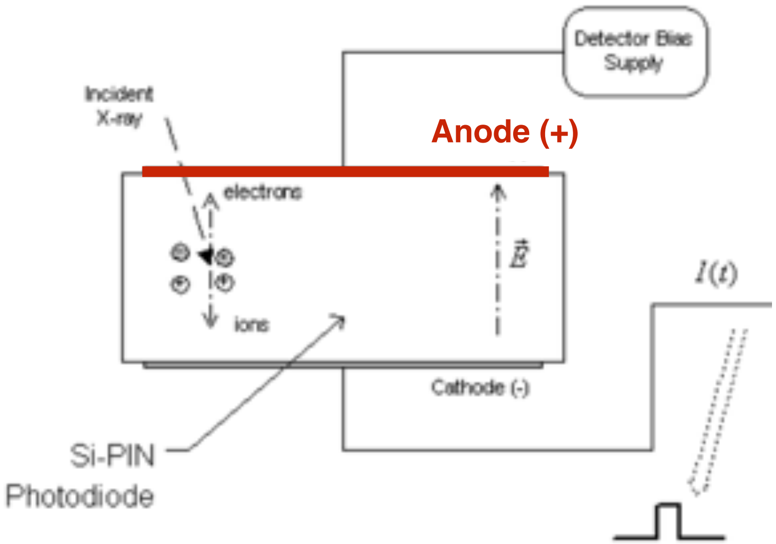
Possible materials: Silicon

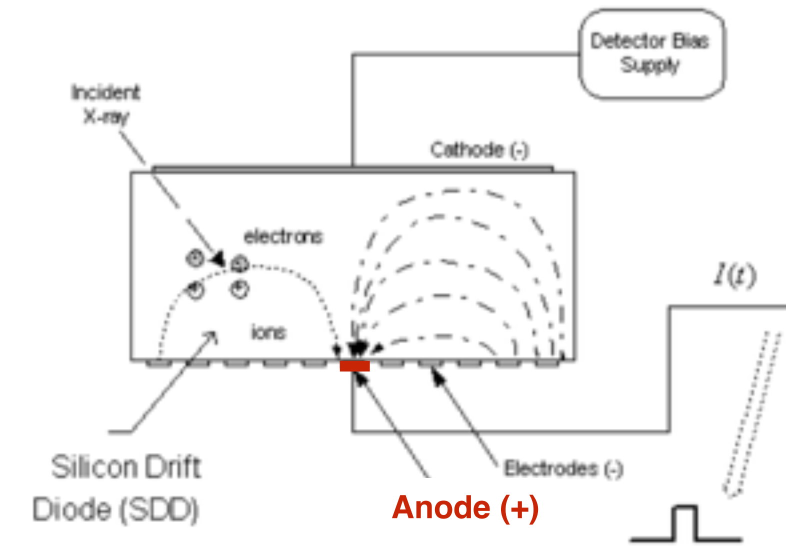
Generally available: CCD, Si-pixels, Si-strip, Si-drift detectors

- CCD (charge-coupled devices) are slow and not very radiation resistant.
- Silicon pixels are fast and have excellent spatial resolution but they are very expensive and the connection to the electronics (bump bonded) is difficult.
- Silicon strip detectors have worse spatial resolution
- Silicon drift detectors
  advantage: very low noise, i.e. high E resolution
  at low photon E



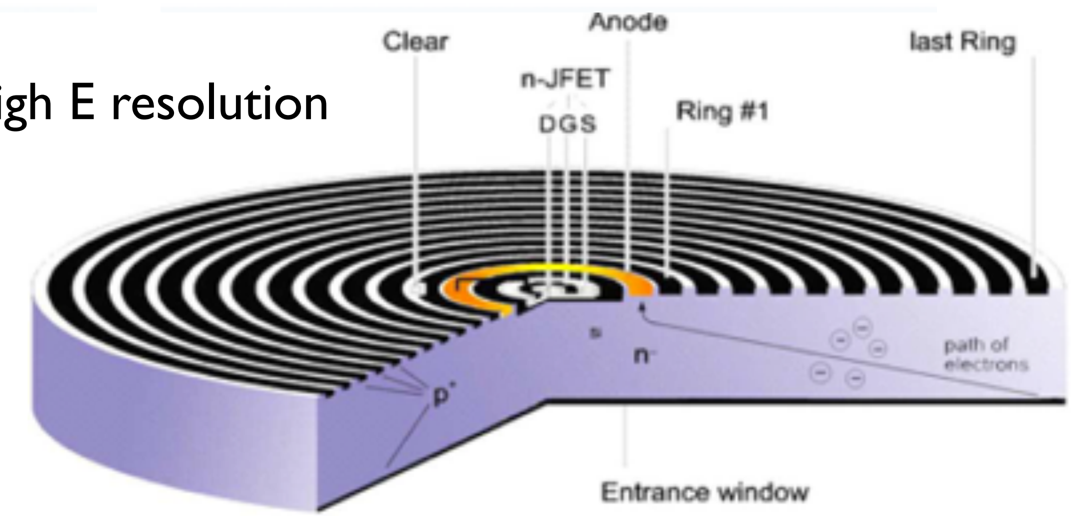
Possible materials: Silicon



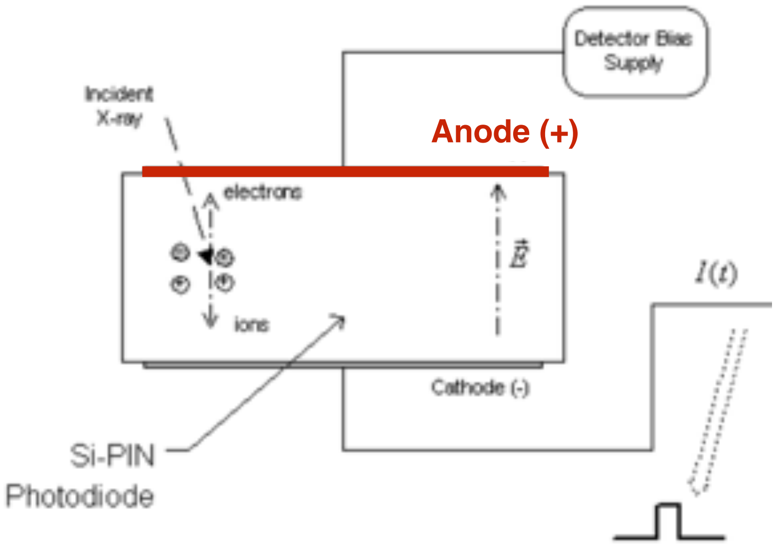


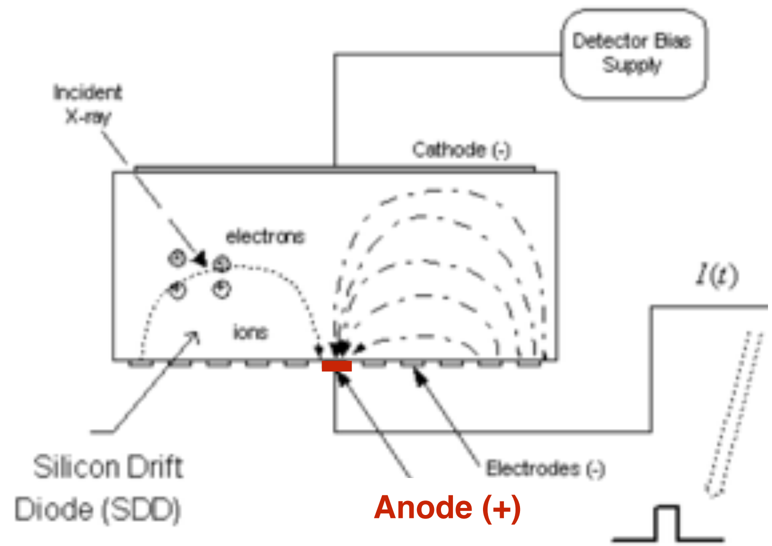
Silicon Drift Detector Application Note
Amptek -X-Ray Detectors and Electronics

- Silicon drift detectors (SDD)
  advantage: very low noise, i.e. high E resolution at low photon E



Possible materials: Silicon





Silicon Drift Detector Application Note
Amptek -X-Ray Detectors and Electronics

- Silicon drift detectors (SDD)
  advantage: very low noise, i.e. high E resolution
  at low photon E

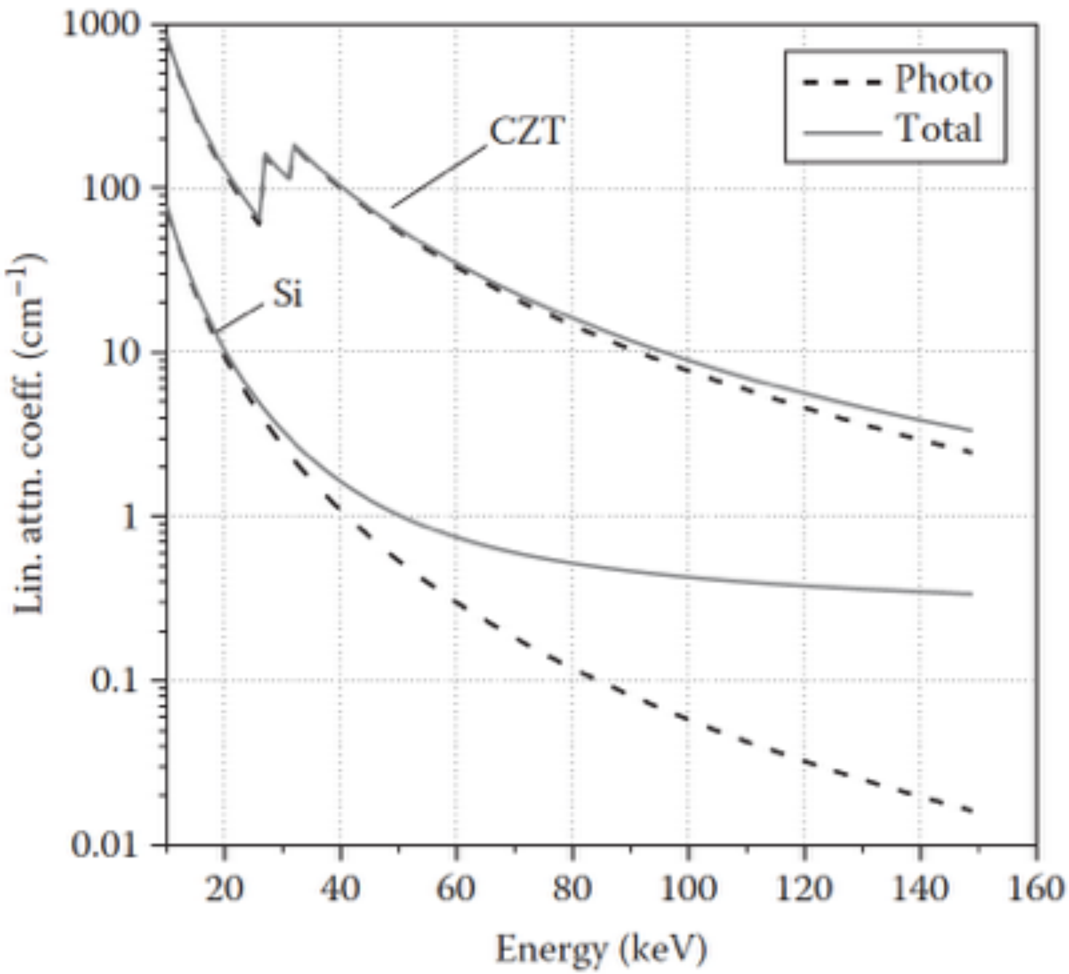
- small anode area in SDD = small capacitance
- active volume is still large (determined by electrodes)
- the dominant noise source is voltage noise proportional to capacitance
- SDD has lower noise than planar Si-PIN, particularly at very short shaping times

Possible materials: CdZnTe

Alternative semiconductors to Si:

- CdZn or CdZnTe crystals
- 100% photon absorption for 60 keV in 3 mm long crystals
- 86% attenuation for 120 keV
- 93% photoelectric fraction at 60 keV and 85% at 120 keV
- carrier creation energy 4.64eV per electron–hole pair

Note: high-purity Ge detectors have best energy resolution carrier creation energy of 2.95 eV and highest charge collection efficiency, but need to operate at liquid nitrogen temperatures

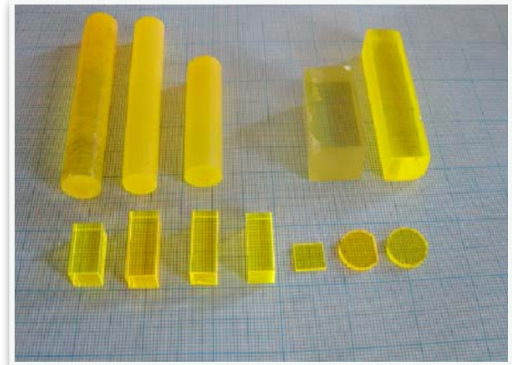


“Radiation Detectors for Medical Imaging”, K. Iniewski, CMOS Emerging Technologies Research Inc., Vancouver, British Columbia, Canada

Possible materials: Scintillators

Required characteristics:

- high light output
- very fast scintillation decay
- and high atomic number (high density)



Y3Al5O12:Ce

⇨ Cerium-doped inorganic scintillators provide these characteristics

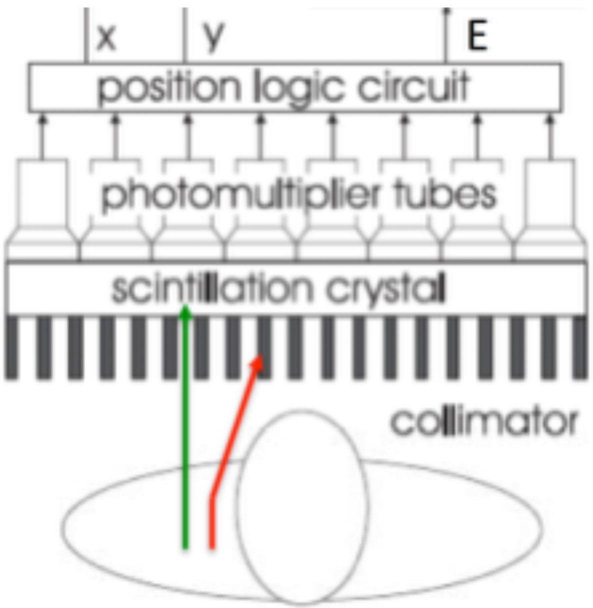
Characteristics of Some Candidate Scintillator Materials for SiPM-Based Indirect Conversion ED-PC X-Ray Imaging Arrays

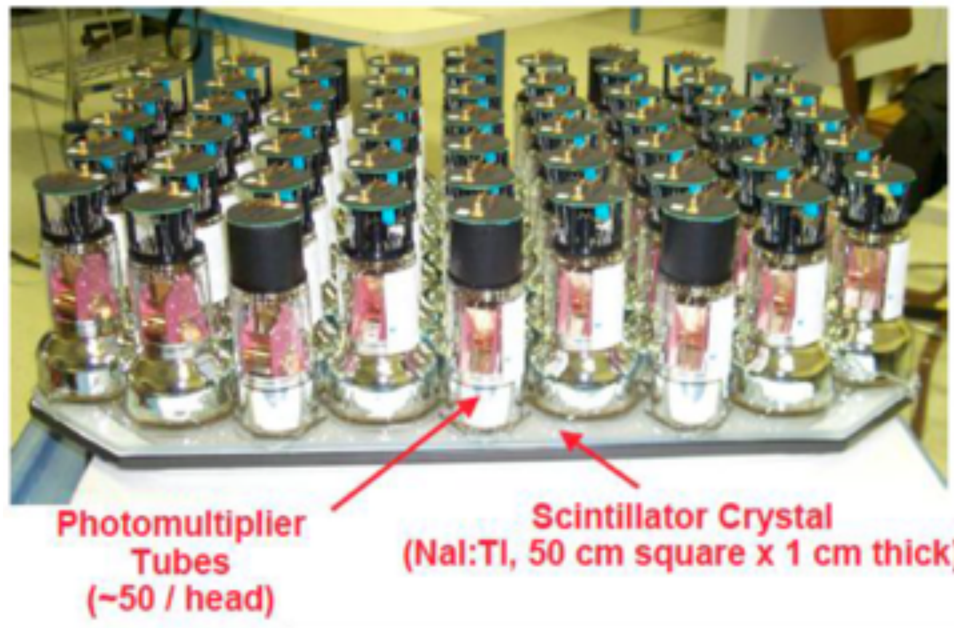
Scintillator	LSO	LYSO	YAP	LuAP	LaBr
Density (g/cc)	7.4	7.1	5.4	8.3	5.3
Light yield (photons/keV)	27	32	21	10	61
Effective, Z	66	64	31.4	65	46.9
Principal decay time (ns)	42	48	25	18	35
Peak wavelength (nm)	420	420	370	365	358
Index of refraction	1.82	1.8	1.94	1.95	1.88
Hygroscopic	No	No	No	No	Yes

lutetium oxyorthosilicate (LSO),
lutetium yttrium oxyorthosilicate (LYSO),
yttrium aluminum perovskite (YAP),
lutetium aluminum perovskite (LuAP),
lanthanum bromide (LaBr3)

SPECT: Anger camera

First detector used in medicine for emission/functional imaging



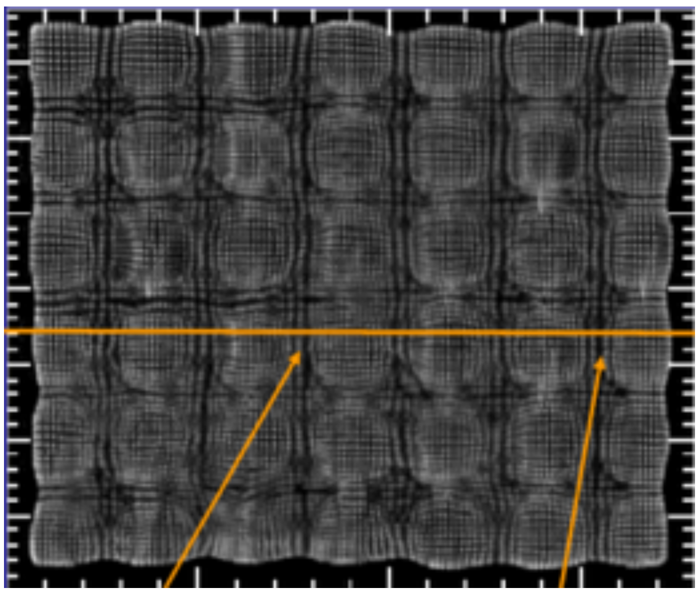


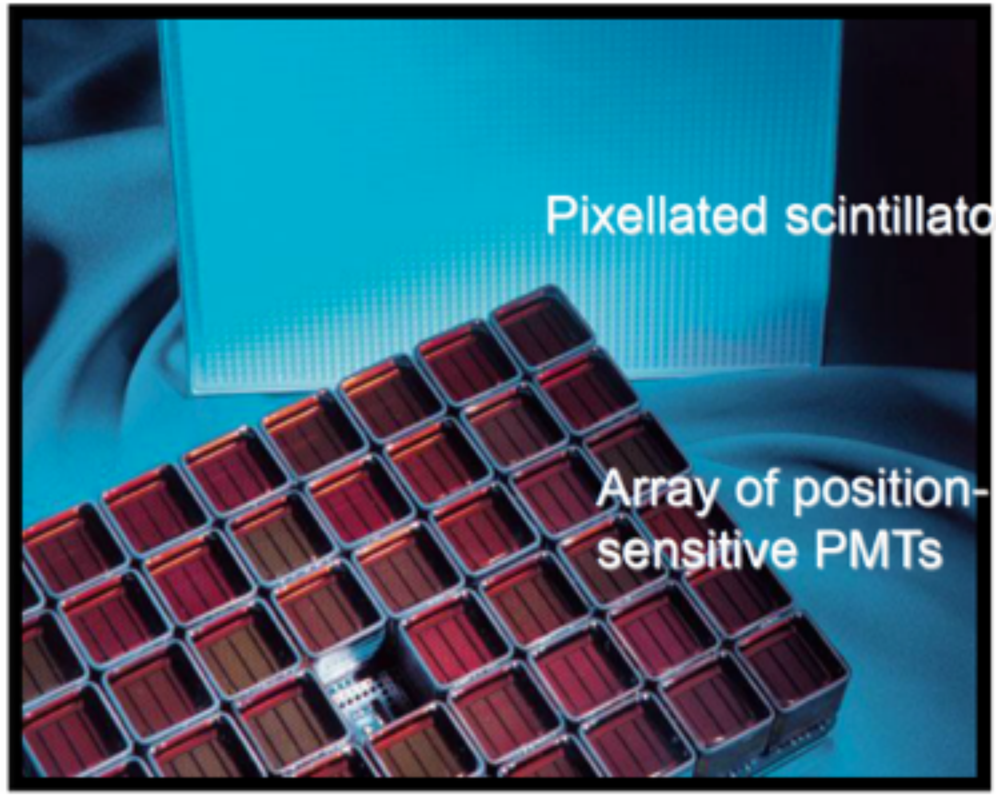
Detection of gamma photons: 100 -200 keV

- > 85% detection efficiency
- good energy resolution (~15%)
- spatial resolution ~ 4mm

SPECT: Anger camera

Evolution in photomultiplier tubes:



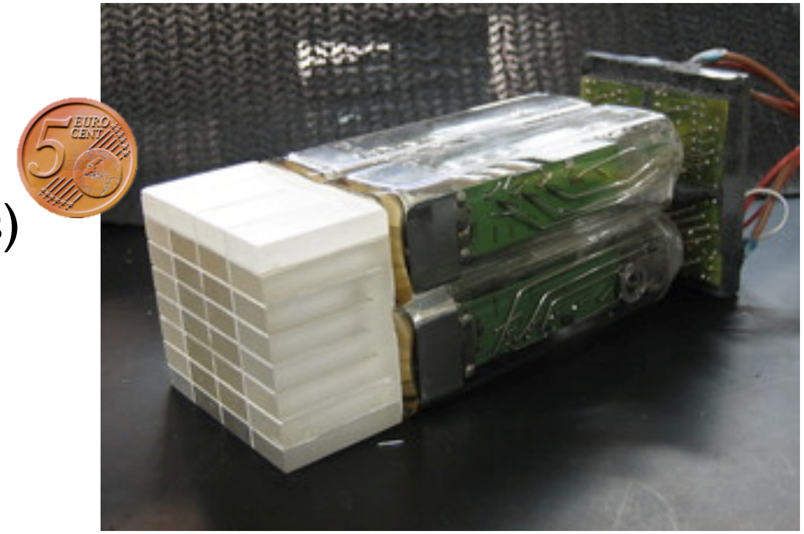


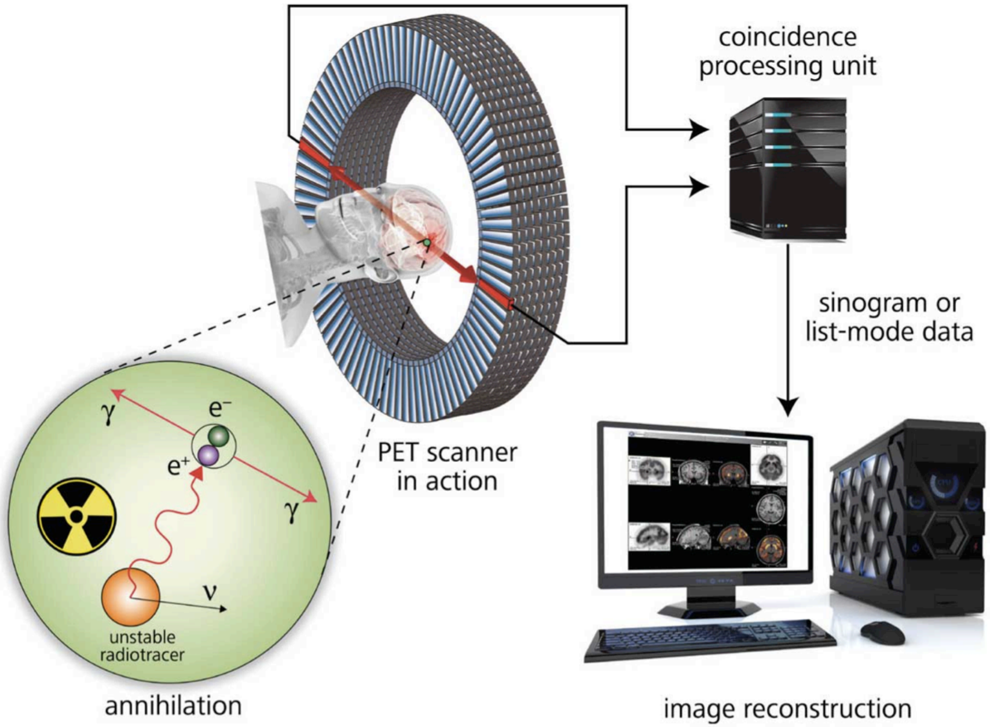
- spatial resolution improved to ~ 2mm using position sensitive PMT and pixellated scintillators

PET: Block detector

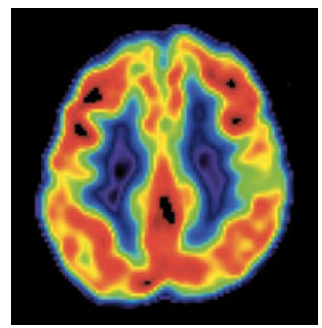
Detection of 2 gamma photons: 511 keV

- > 85% detection efficiency *
- good energy resolution (~15%)
- spatial resolution ~ 4 mm (clinical scanners)
- time resolution for background rejection





Brain slice

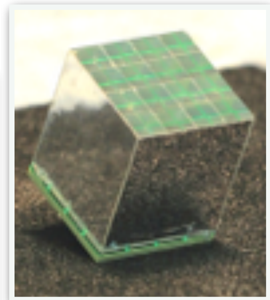


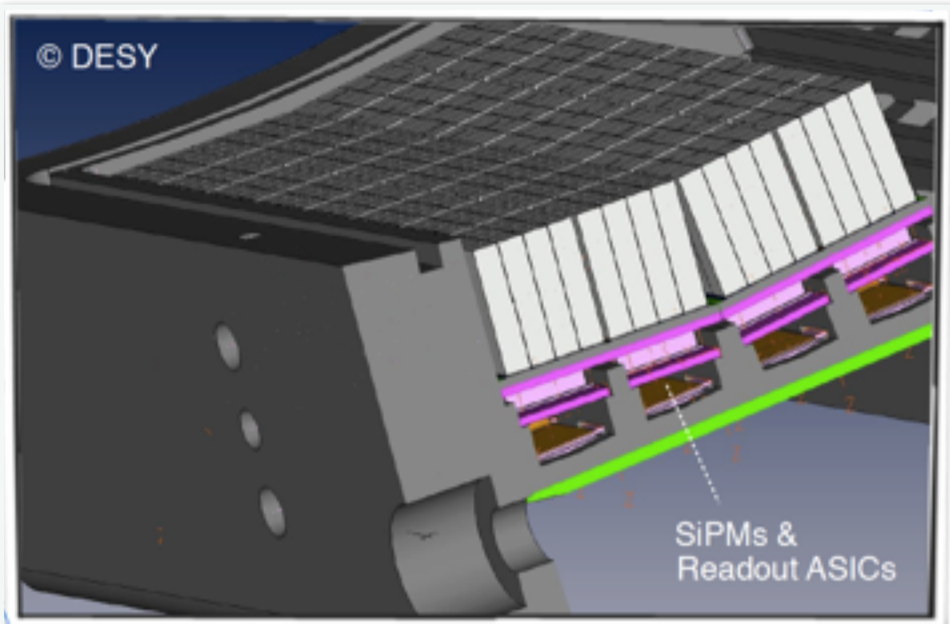
* Factor ~3 higher E than SPECT: need either longer crystals (NaI) or denser materials (BGO, LSO, or techniques)
e.g. μ_{BGO}~ 3 x μ_{NaI} (μ_{BGO} = 0.96 cm^{-1}): Typical BGO length for PET ~ 3cm

PET: State of the art

Intense R&D in animal PET detectors
& in detectors for personalised medicine, i.e.
organ specific multi-modal systems (mammography, pancreas/prostate endoscopy, neurology, …)

- spatial resolution ~1.2-2 mm (physics lim. ~0.8 mm)
- time resolution ~ 200 ps (push to 10 ps ???)

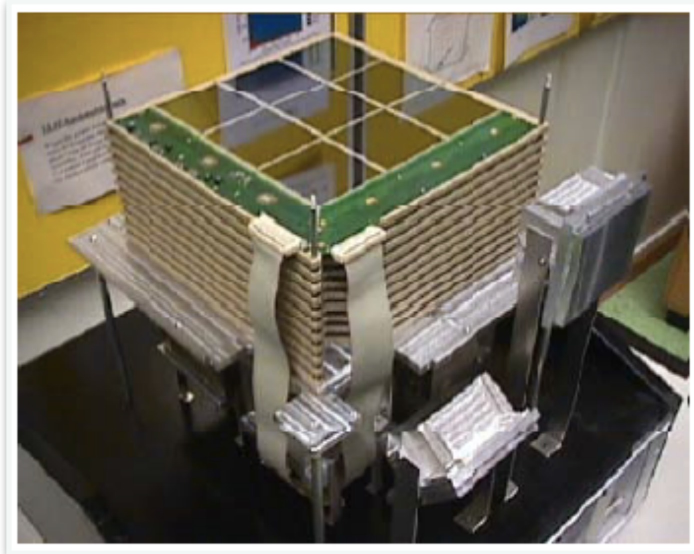




Direct coupling of crystals to silicon-based photo-detectors (SiPMs) allows ~1x1mm^2 single channels

Digital SiPM offers even smaller single ch. The limit is in the diameter of crystal fibres or back to block detector

PET: alternative technologies



Current R&D on:

1. Silicon strip detectors
   much higher spatial resolution… but multi-layer structure needed to absorb photons

2. Heavier semiconductors (Ge, CdZnTe, …)
   Expensive, less commercially available, need R&D

3. Liquid noble gases (LXe)
   Complex operation, require vacuum cryostat

Both semiconductors and gases are slower than scintillators (no ToF)

Clinical PET instrumentation seems to be scintillator oriented also for the future
Solid state detectors could be a valid alternative but need R&D.

Integrating detectors

- First transmission imaging used collimated isotope sources & Anger cameras
- X-ray tubes replaced sources and provided high photon outputs and small spot sizes

Problem: the photon-counting scintillation detectors could not handle the high count rates from X-ray tubes

Solution: use scintillation detectors in energy integration mode, i.e. integral of output signal pulses is proportional to photon flux

Consequence:

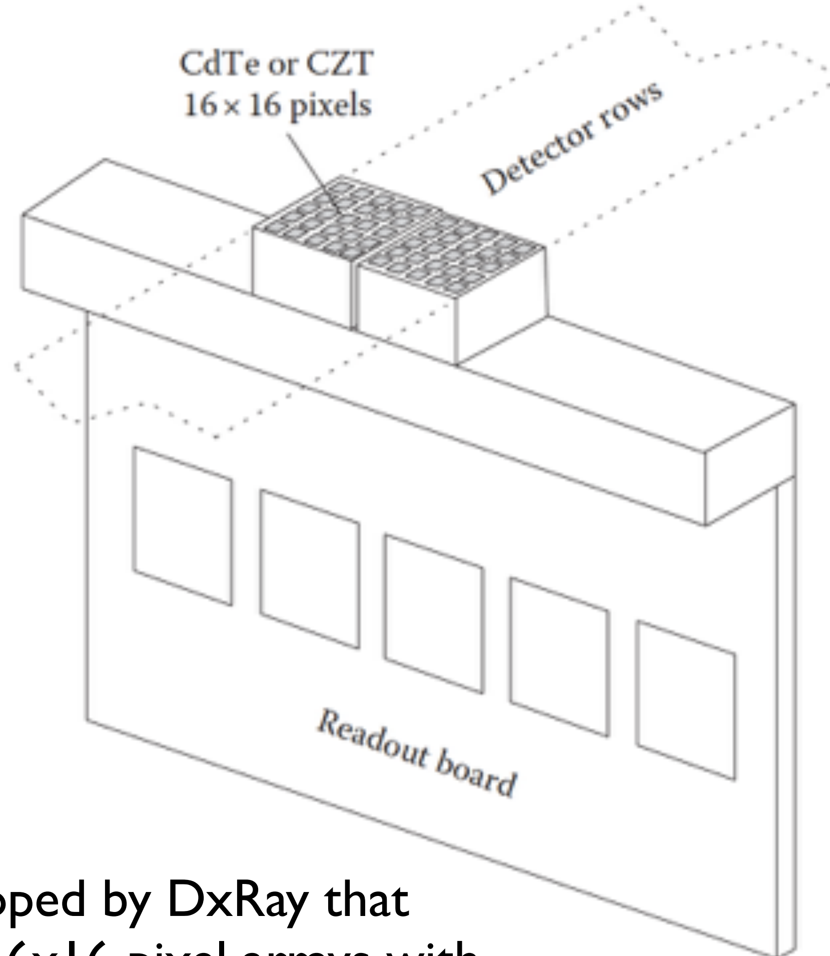
- photon-counting detectors abandoned till 1990
- in energy-integrating detectors statistical separation of signal and noise not possible
- SNR linked to image contrast

To improve contrast in X-ray and CT detectors (or reduce patient exposure) need to reduce detector electronic noise ⇨ re-birth of photon-counting detectors

Photon-counting detectors

R&D on pixilated photon-counting detector mainly focused on semiconductors:
high-purity Ge, CdTe, CZT, and Si

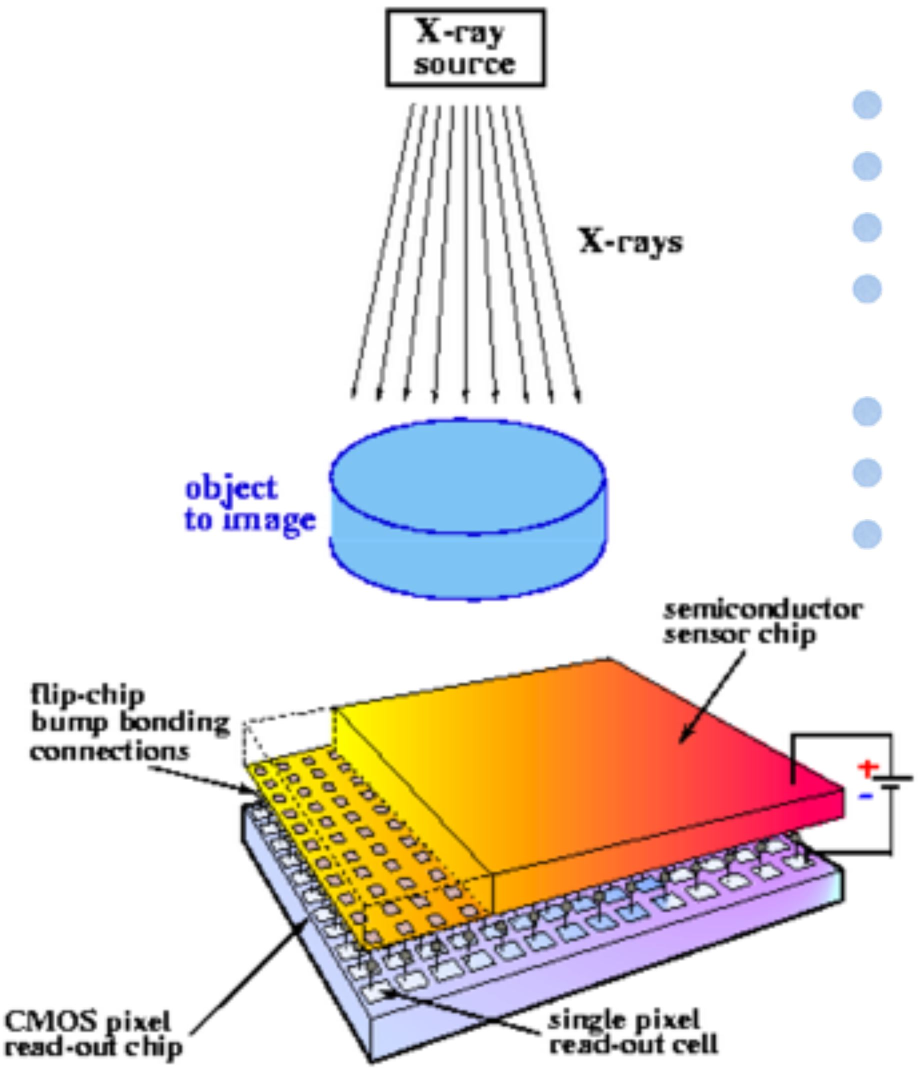
Alternatives: micro-channel plates, gas-filled detectors



“Radiation Detectors for Medical Imaging”, K. Iniewski, CMOS Emerging Technologies Research Inc., Vancouver, British Columbia, Canada

Photon-counting detector module developed by DxRay that includes 2 CdTe (or CZT) crystals with 16x16 pixel arrays with 1x1 mm^2 pixel size. Many modules can be tiled up to extend the detector rows to a clinically applicable level.

Photon-counting detectors



Medipix chip family:

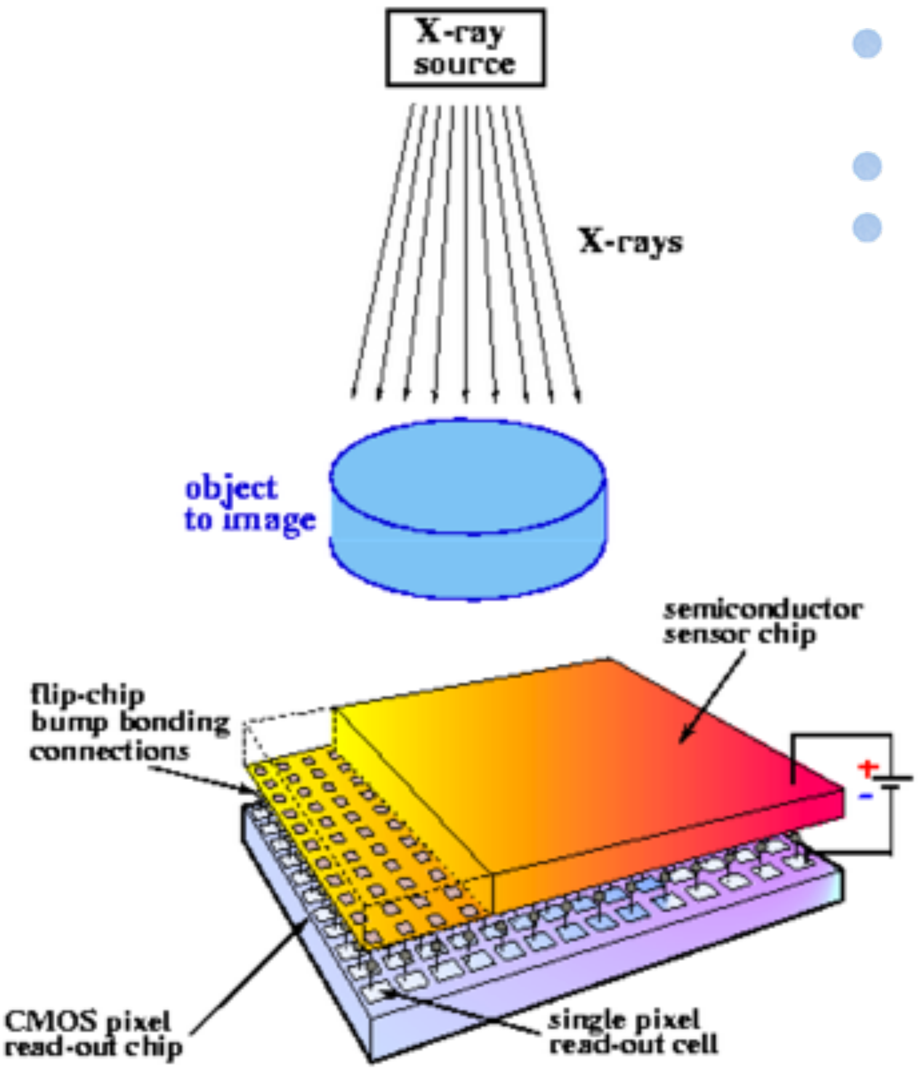
- CMOS chip for pixel detectors (LHC driven)
- bump bond to segmented semiconductors
- images based on the number of hits/frame
- can image X-ray photon (1keV < E < 20keV)

- square pixel size of 55 μm side length
- chip area of about 2 cm^2 / 256 x 256 pixels
- rate: 100 Kcount/pixel/s or 16 Mcount/mm^2 /s

Limitation:

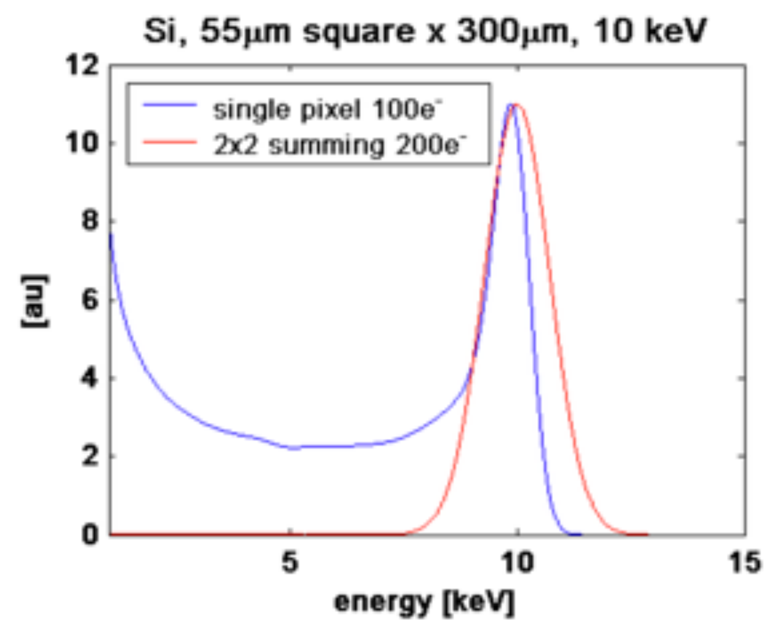
- small sensitive area ⇨ N x 2.8 cm
- difficult scaling to clinical imaging requirements, i.e. ~ 25 cm^2 for mammography

Photon-counting detectors



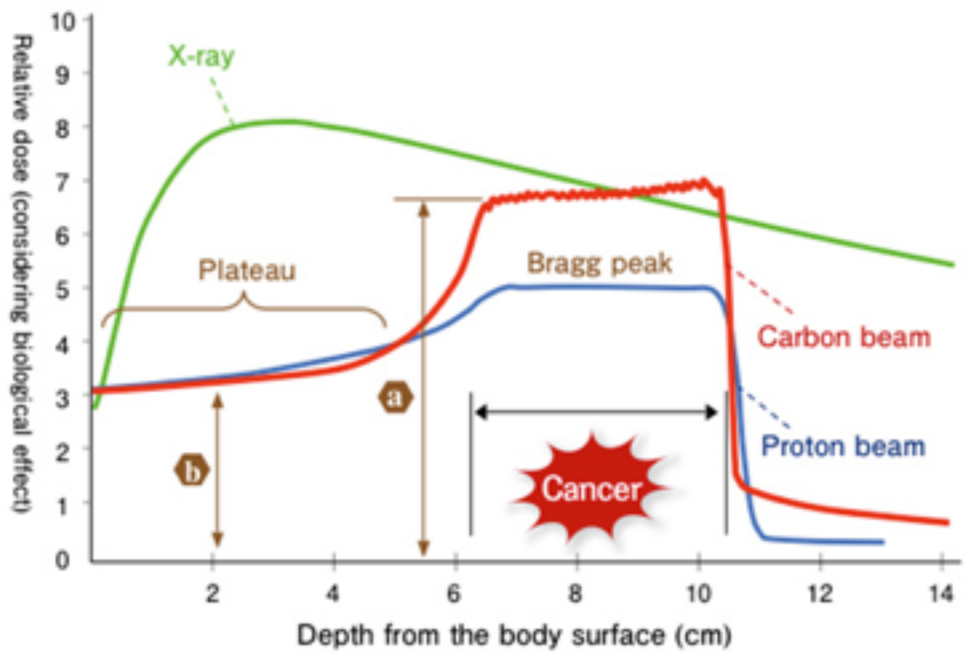
Medipix3 - correction of charge sharing:

- due to diffusion charge is collected in pixels neighbour to the one where the hit happened
- distorted energy spectrum of single pixel
- mitigate the effect of charge sharing by summing charge between neighbouring pixels and allocating the sum or hit to the individual pixel with the highest collected charge.



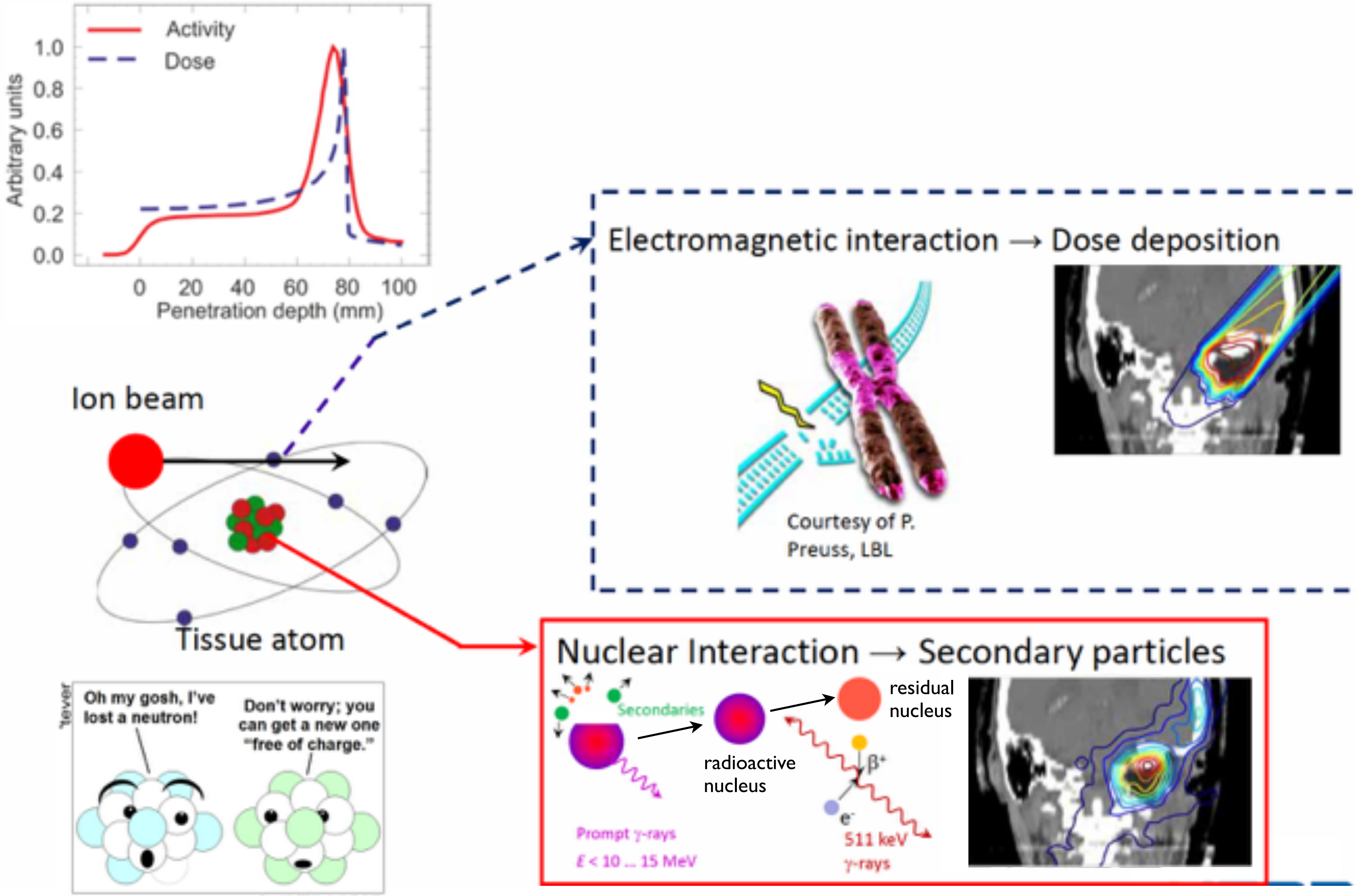
Particles against cancer

- Advantages of hadrontherapy for localised treatment of tumours :
  - More localised energy deposition in target due to the Bragg peak
  - Better biological efficiency of hadrons compared to photons (LET)



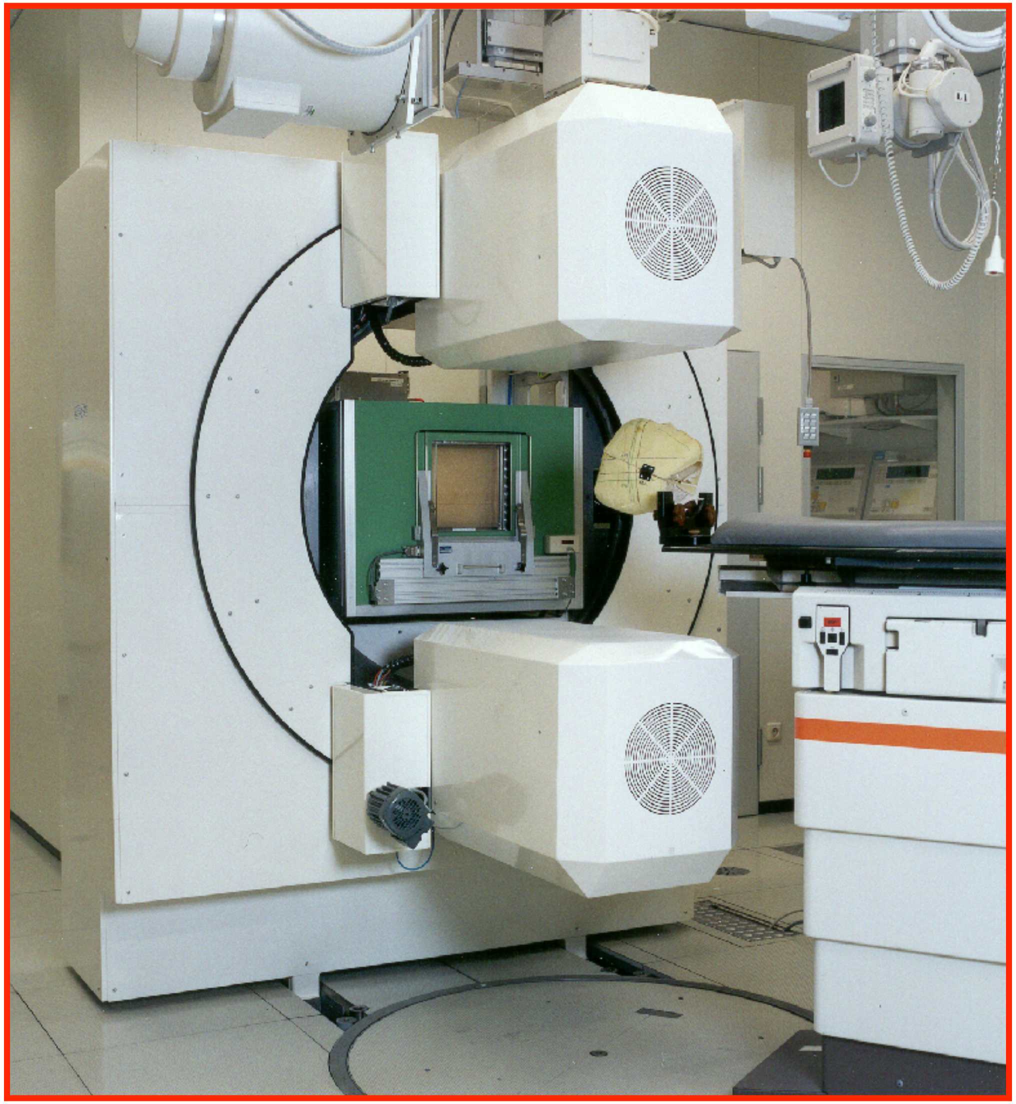
- Novel Imaging Systems for in vivo Monitoring and Quality Control during Tumour Ion Beam Therapy (proton, carbon)

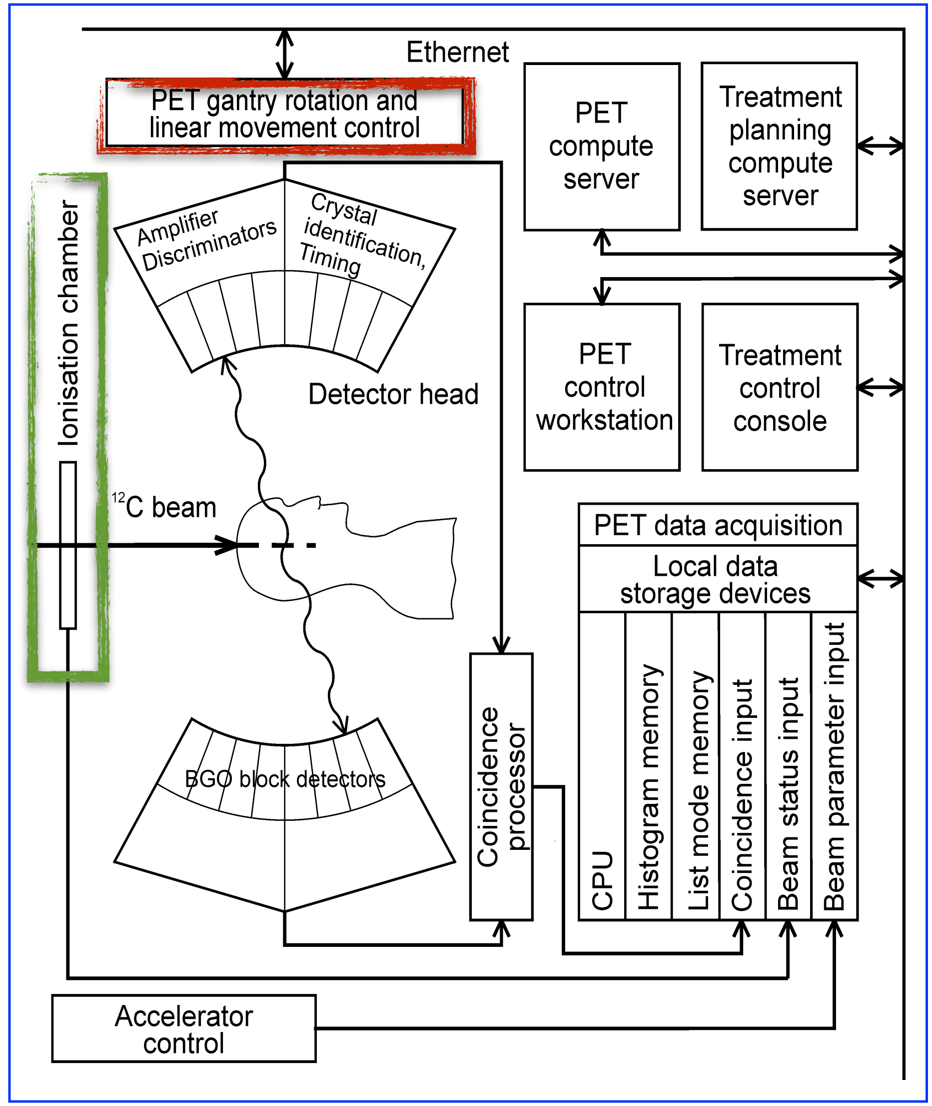
Particles against cancer



Detectors for radio-therapy

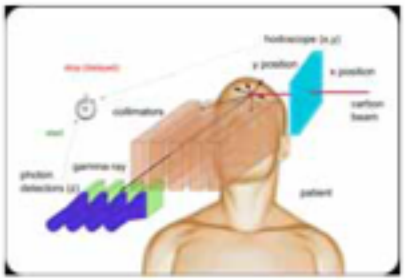
In-beam PET: ^{12}C-therapy at GSI Darmstadt





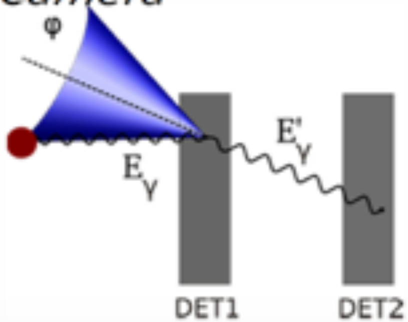
Detectors for radio-therapy

Collimated γ-Kamera



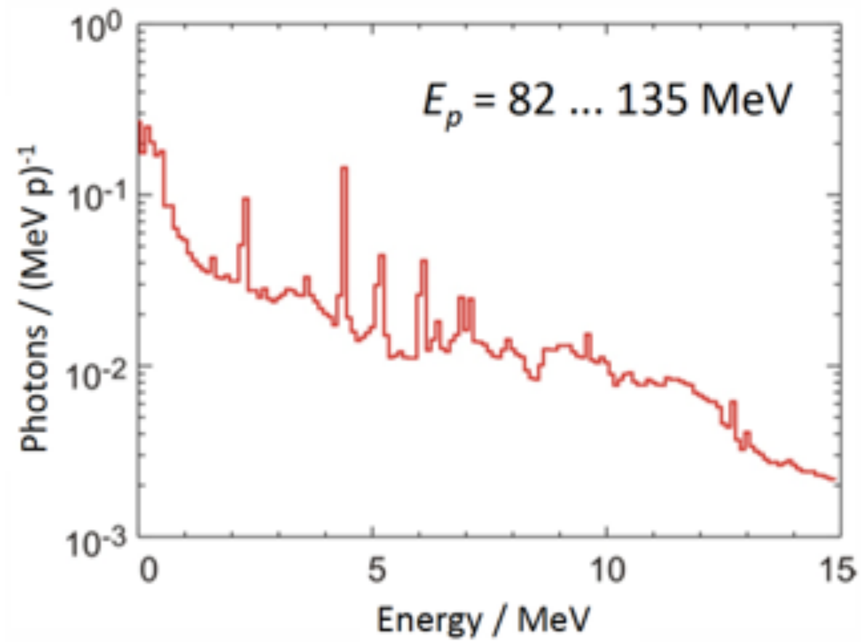
Slit camera

Compton Camera



Spectrum of prompt gammas:

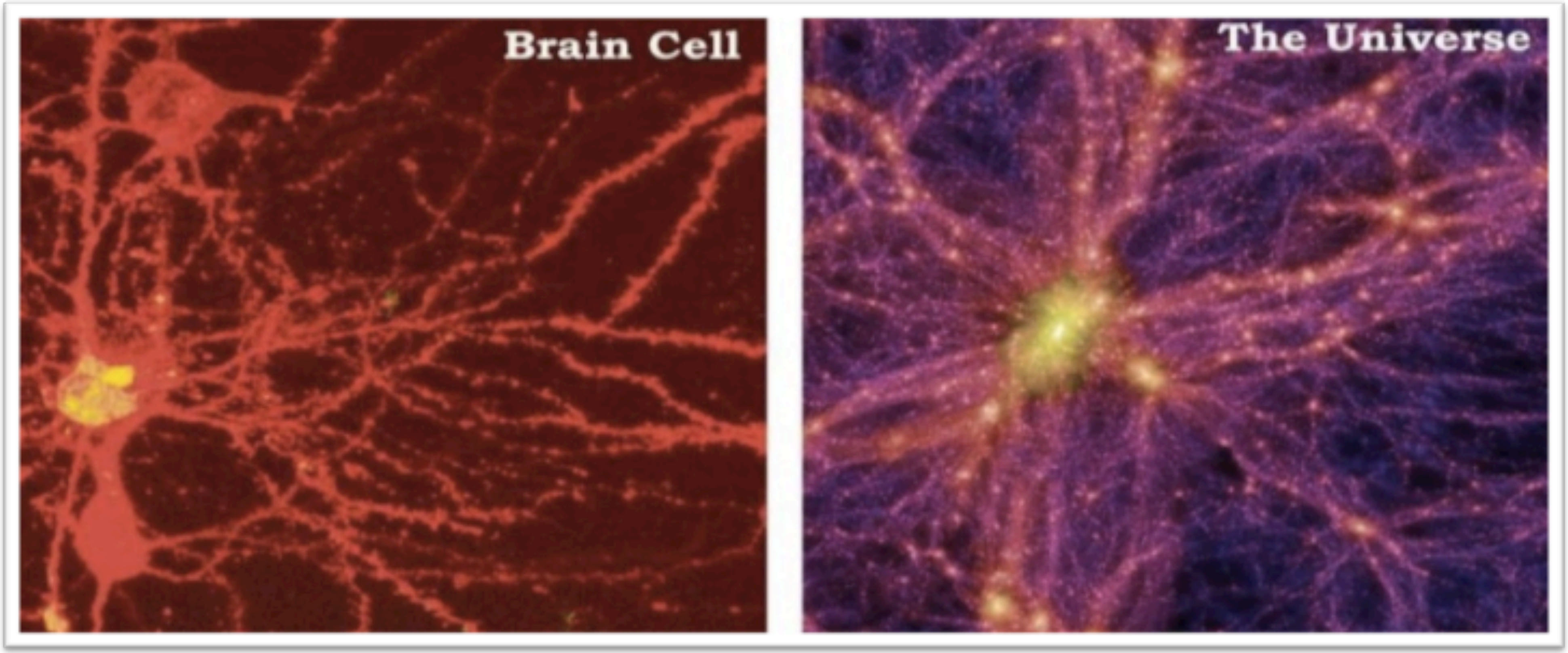
- approximately 10^9 gammas per Gy
- photon energy: 0 … >10 MeV



- Anger camera optimised for ~100-200 keV

New detector concepts needed !!

THANK YOU FOR YOUR ATTENTION !



Mark Miller

Virgo Consortium

Backup

Comparison of the performances

Imaging modality	Type of imaging	Examination time	Spatial resolution
PET	Functional (picomolar sensitivity)	30-45’	4-6 mm
SPECT	Functional	30’	6-8 mm
MRI	Anatomical Functional (millimolar sensitivity)	10’	0.5 mm
CT	Anatomical	1’	0.5 mm