Joint CI-JAI advanced accelerator lecture series

Imaging and detectors for medical physics

Lecture 2: Detectors for medical imaging

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Course layout

Day	AM 09.30 – 11.00	PM 15.30 – 17.00
Week 1		
6th June	Lecture 1: Introduction to medical imaging	Lecture 2: Detectors for medical imaging
7th June	Lecture 3: X-ray imaging	
8th June		Tutorial
Week 2		
13th June	Lecture 4: Radionuclides	
14th June	Lecture 5: Gamma cameras	Lecture 6: SPECT
16th June	Lecture 7: PET	
Week 3		
22nd June	Tutorial	

Books

1. N Barrie Smith & A Webb
   Introduction to Medical Imaging
   Cambridge University Press

2. Edited by M A Flower
   Webb's Physics of Medical Imaging
   CRC Press

3. A Del Guerra
   Ionizing Radiation Detectors for Medical Imaging
   World Scientific

4. W R Leo
   Techniques for Nuclear and Particle Physics Experiments
   Springer-Verlag

From particle physics to medical imaging

*"The significant advances achieved during the last decades in material properties, **detector** characteristics and high-quality electronic system played an ever-expanding role in different areas of science, such as high energy, nuclear physics and astrophysics. And had a reflective impact on the development and rapid progress of radiation detector technologies used in medical imaging."* **D. G. Darambara, Nucl. Inst. And Meth. A 569 (2006) 153-158**

- Nuclear medicine: fertile field for applications of technology from high energy physics
- Main goal: better image quality for same dose delivered or reduce dose delivered for same image quality

Analogue –vs– digital

Analogue imaging (films)	Digital imaging
Continuous range of possible optical densities up to some limiting value	Discrete and limited range of optical densities
Narrow exposure latitude → strict exposure requirements	Very wide exposure latitude
Little possibility of image processing	Image processing possible + needed to overcome limitations of manufacturing processes (bad pixels, spatial sensitivity variation)
Only one image display	Various image displays possible
Cheap but one use	Expensive but multiple use
High resolution	Low(er) resolution[1]

[1]Not clinically significant when choosing right matrix size and image receptor to match the application

Analogue → digital imaging (= more quantitative information available) thanks to:

1. Integrated electronics
2. Fast computers

Specific requirements for medical imaging detectors

- Detector for medical application = special detector with its own specifications:

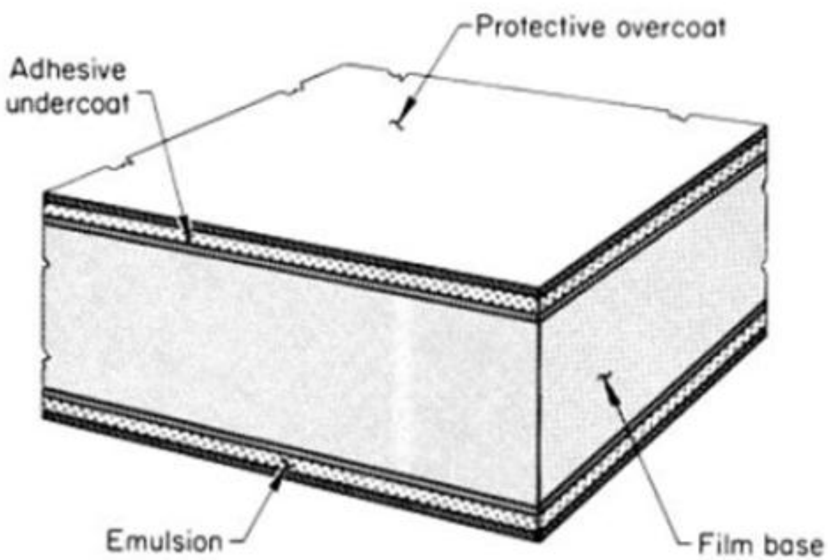
Detection range	− Low energies: 18 keV for mammograms for ex.
Read-out	− No trigger (no bunch crossing) → self-triggering electronics or free running − High acquisition rates > GHz − Manageable number of read-out channels
Event size	− Can be small 1 bit ÷ 10 bytes
Geometry	− Large area often required − Almost no dead space
Patient's requirements	− Meet stringent ethical requirements and regulation − Ensure patient's comfort
Market	− Can be large: $10^3 \div 10^6$ units

Digital X-ray radiology

- Primary requirements for digital X-ray detector for radiology:
  1. Speed
  2. Area
  3. Spatial resolution
  4. Dose rate to exposure ratio
- Trade-off between:
  1. Spatial resolution & dynamic range
  2. Read-out speed & time resolution
  3. Contrast & exposure time

Radiographic films

Ref. 3 – Chapter 2.3



- Photographic emulsion = mixture of silver halide grains of different size + gelatine
- Emulsion deposited on transparent support = base
- Emulsion coated with thin protective layer
- Limitations:
  1. Limited dynamic range
  2. Lack of digital processing

Image formation

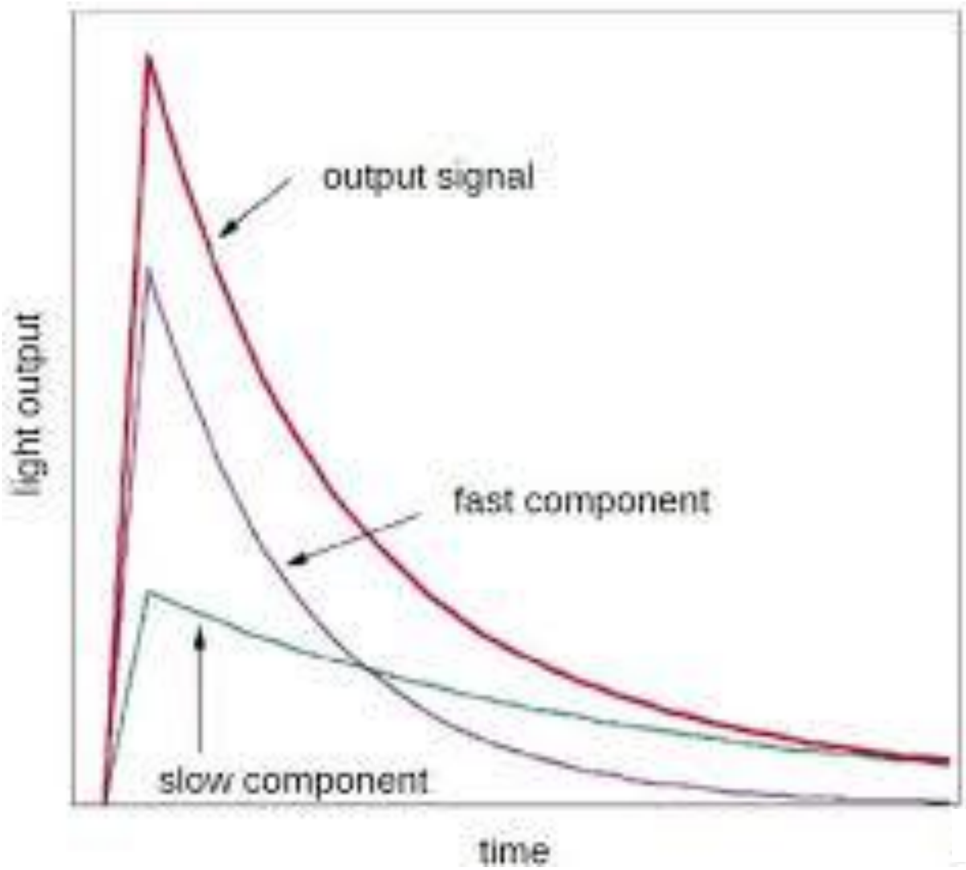
- X-ray arriving on film interact with silver halide crystals $\rightarrow e^-$ released
- e^- migrates until trapped in a grain where neutralises silver ion
- If about four silver ions are captured and neutralised at same grain $\rightarrow$ stable configuration reached = grain said to be sensitised
- Development of film = sensitised grains are reduced to metallic silver + unsensitised grains are removed
- 1 X-ray sensitises one or more grains, while optical photons sensitise only one grain

Scintillators

Ref. 4 – Chapter 7

- Scintillation = emission of light generated by the passage of a particle / radiation
- Radiation loses energy = photoelectric + Compton interactions $\longrightarrow$ excitation of atoms and molecules $\longrightarrow$ de-excitation = emission of optical photons:
  1. Fluorescence: reemission occurs within 10^{-8} s $\cong$ time for atomic transition
  2. Phosphorescence or afterglow: excited state metastable $\longrightarrow$ reemission delayed by few μs to hours depending on scintillator material

Scintillation light



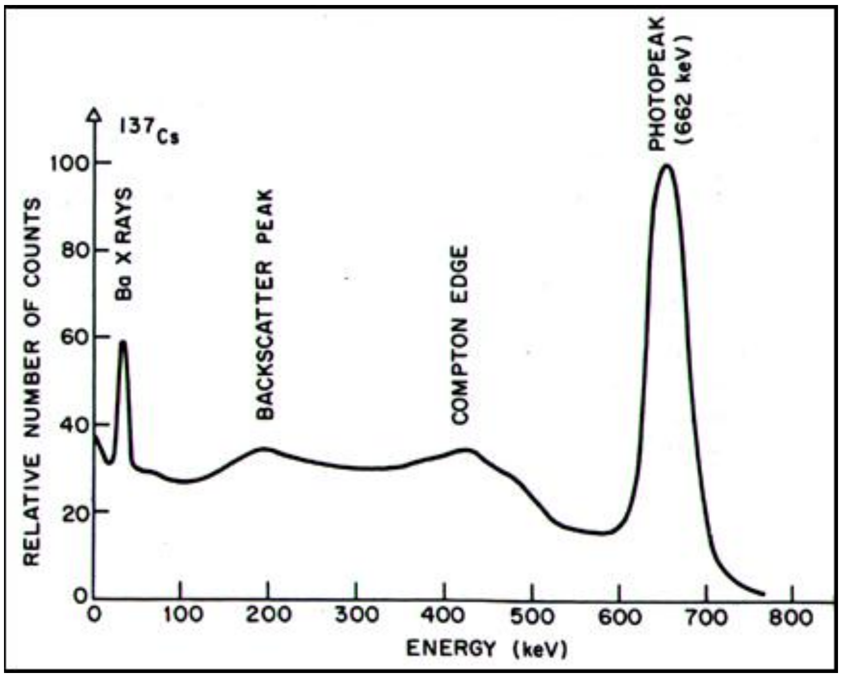
$$N = A\exp\left(\frac{-t}{\tau_f}\right) + B\exp\left(\frac{-t}{\tau_s}\right)$$

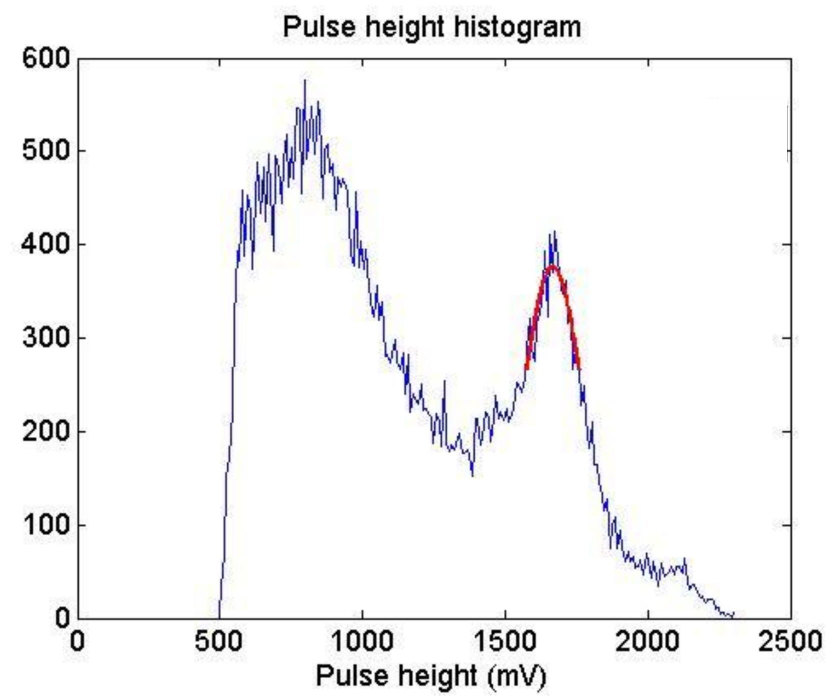
$N =$ n of photons emitted at time t

$\tau_f, \tau_s =$ decay constants of fast and slow components

$A, B =$ relative amplitudes of fast and slow components

Example output signal: ^{22}Na source





FP7 funded ENVISION project – University of Oxford
With help from University of Sussex

Scintillators

	Inorganic	Organic
Material	Mainly alkali halides with small activator impurity	Aromatic hydrocarbon compounds with benzene-ring structures
Density	High	Low
Atomic number	High	Low
Stopping power	High	Low
Light output	High	Low
Energy resolution	High	Low
N of photons generated	Linear[1]	Non linear[1]
Decay time	~500 ns	Few ns or less
Temperature dependence	Yes	No
Hygroscopic	Usually yes	No

[1]With energy of incident radiation

Commonly used inorganic scintillators

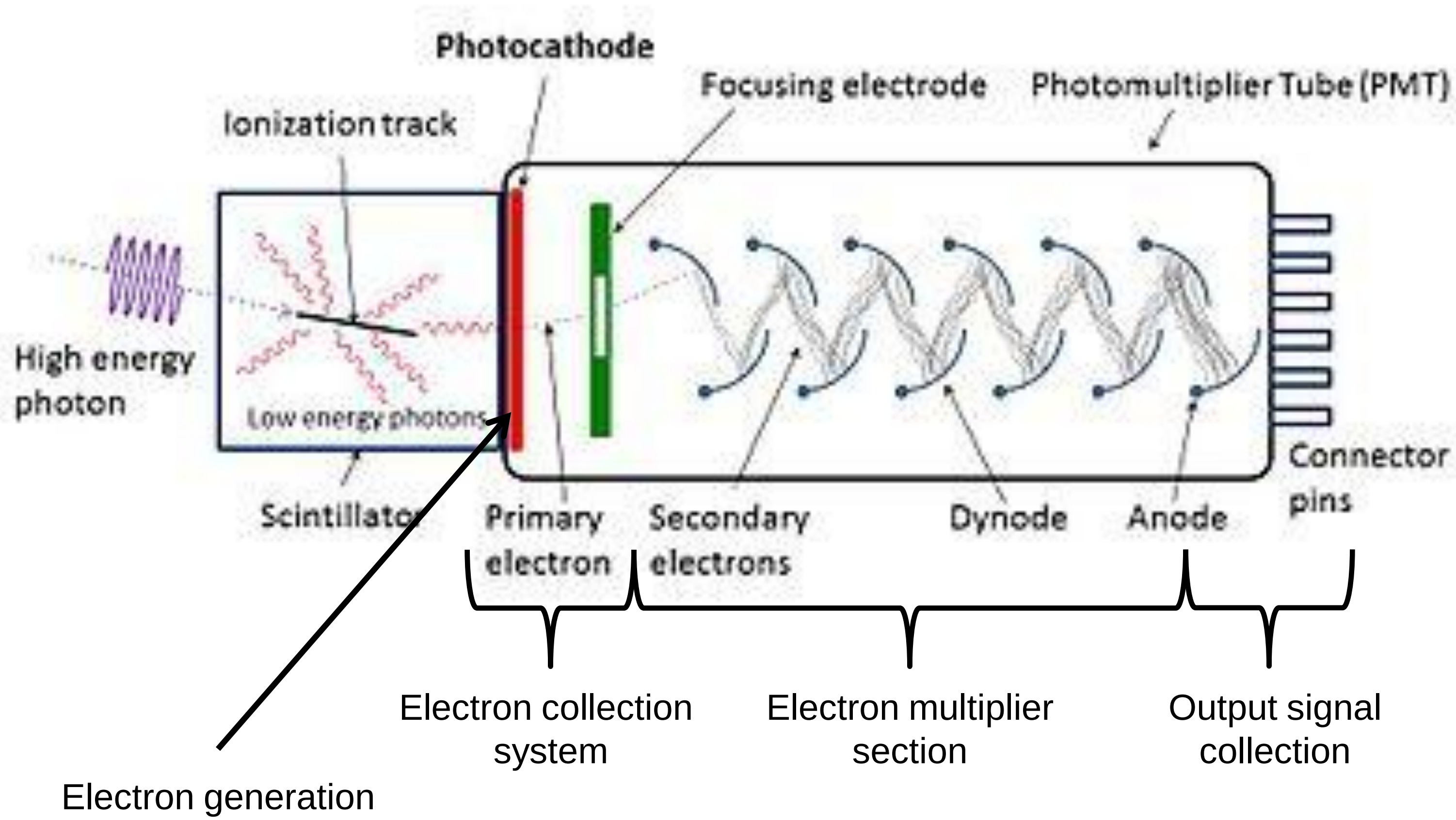
	NaI(Tl)	Ba_2F	BGO	GSO	LSO	LYSO	$LaBr_3$
Density (g cc^{-1})	3.67	4.89	7.13	6.71	7.4	7.3	5.1
Effective atomic number	51	54	75	58	66	66	~50
Relative light yield	100	5 16	15	20	75	75	166
Light decay time (ns)	230	0.6 620	300	~60	~40	40	16÷30
Emission wavelength (nm)	410	195÷220 310	480	430	420	428	380
Refractive index	1.85	1.49	2.15	1.85	1.82	1.82	~1.9
Attenuation length (mm) at 511 keV	25.6	22.7	11.2	15.0	11.4	11.6	21.3

Photodetectors

- Convert light into electric signal
- Various types:
  - Photomultiplier tubes (PMTs)
  - Photodiodes (PD) / Avalanche Photodiodes (APDs)
  - Silicon Photomultipliers (SiPMs)
  - ~~Vacuum Phototriodes (VPTs)~~

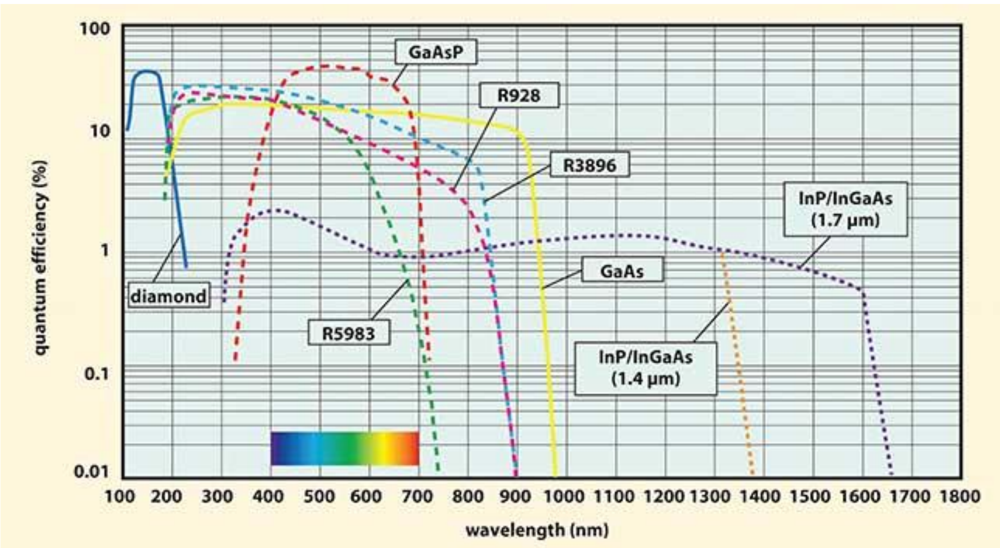
Photomultipliers tubes

Ref. 4 – Chapter 8



PMT: the various stages

Photocathode

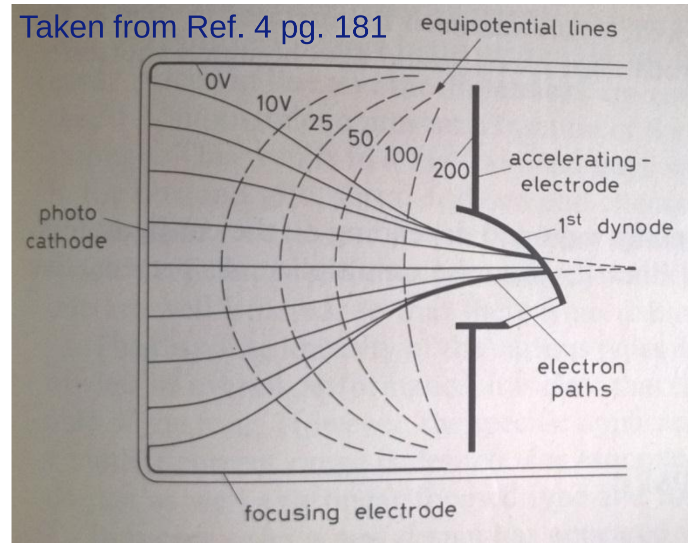


- Quantum efficiency $\eta(\lambda)$:

$$\eta(\lambda) = \frac{N_{pe}}{N_\gamma}$$

- Typical η are $10\%\div40\%$ for λ between $200\div800$ nm

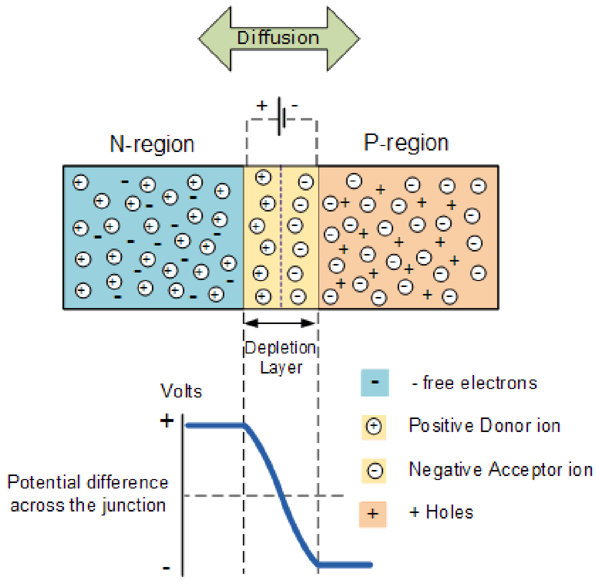
Electron collection system

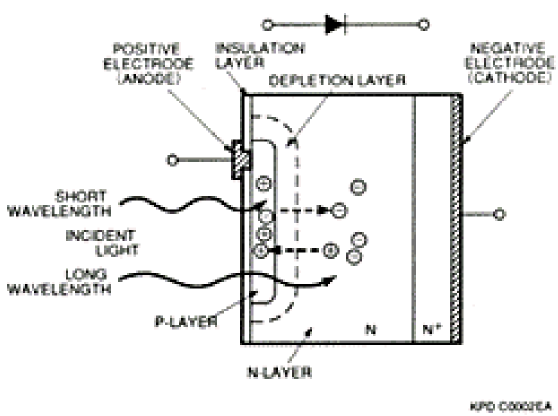


Electron multiplier section

- Conventional PMTs $10\div14$ stages = gain up to 10^7
- Dynodes kept at increasingly higher V

Photodiodes (PDs)





- p–n junction in a silicon substrate wafer
- Incident light creates e–h pairs in the junction
- No intrinsic amplification → small output signal that requires post amplification →
  - Better for applications with large number of optical photons produced → no amplification
  - Amplification introduces noise, especially if fast signal required

Avalanche Photodiodes (APDs)

- APD = PD operated near breakdown voltage = higher operating bias $100 \div 200$ V
- High internal electric field $\rightarrow$ internal avalanche multiplication effect:
  - e^- generated is accelerated by high electric field $\rightarrow$ creates free carriers by impact ionisation
- Internal gain = bigger output signal:
  1. Gain usually 10^2
  2. Gain of 10^3 can be achieved using special doping techniques but requires very high voltage ~ 1500 V

PDs –vs– APDs

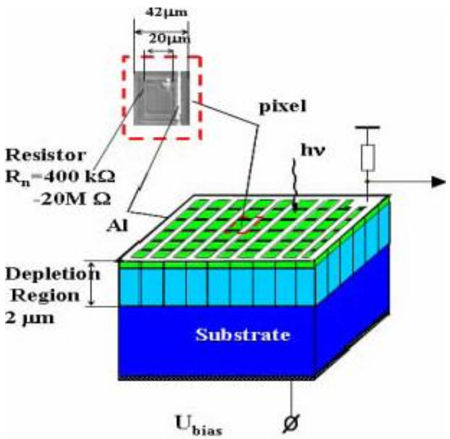
	PDs	APDs
Output signal	Small	Large
Spectral sensitivity	Red/green	Red/green
Response time	Fast	Fast
Bias	$10 \div 100$ V	$100 \div 200$ V ~ 1500 V for special APDs[1]
Gain	$G_{PD} = 1$	$G_{APD} = 10^2 \times G_{PD}$ @ $100 \div 200$ V $G_{APD} = 10^3 \times G_{PD}$ @ ~ 1500 V
Sensitivity to temperature	Yes	Yes
Area	Few mm across	Few mm across
Cost	Cheap	More expensive

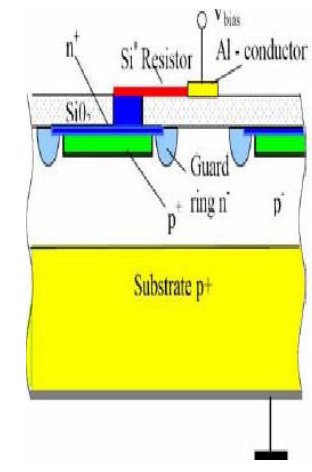
[1]For APDs doped with special doping techniques to obtain very high gain

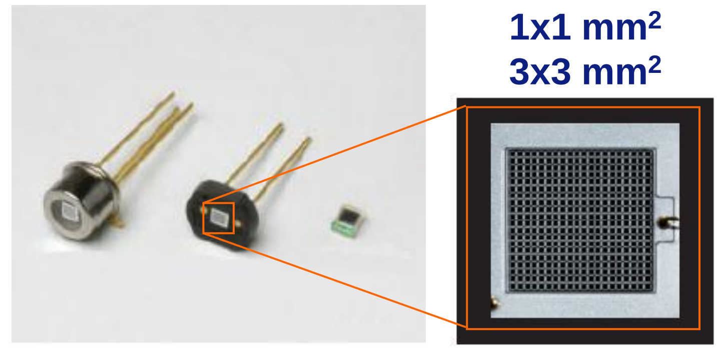
$$Signal_{PD}(1), Signal_{APD}(10^2 \div 10^3) < Signal_{PMT}(10^5 \div 10^8)$$

$\rightarrow$ Limited application to nuclear imaging so far

SiPMs







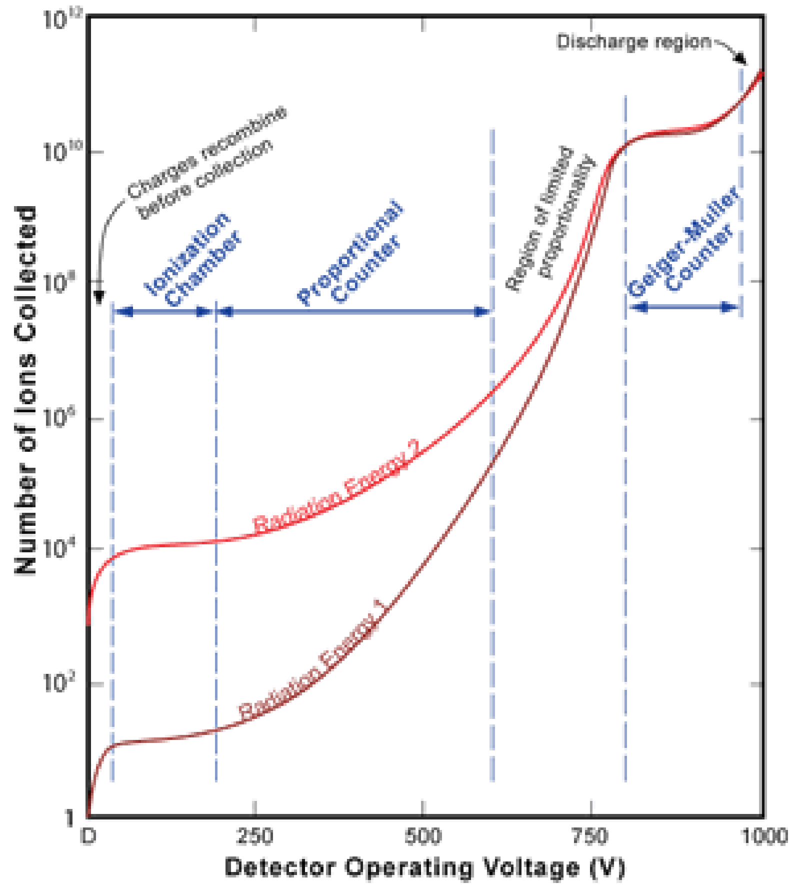
Hamamatsu Inc.

- Array of Silicon Photodiodes on common substrate each operating in Geiger mode
- SiPMs have high speed (sub ns) and gain (10^6) at low bias (< 100 V) and work in high magnetic fields (7 T)

Gaseous detectors

Ref. 4 – Chapter 6

- Volume filled with gas mixture with applied electric field
- Electron – ion pairs created
- Number of pairs collected depends on voltage applied → various regions:
  1. Recombination
  2. Ionisation
  3. Proportional
  4. Limited proportionality
  5. Geiger-Muller



Gaseous detectors in medical imaging

Ref. 2 – Chapter 5.3.5 and Ref. 3 – Chapter 6.3

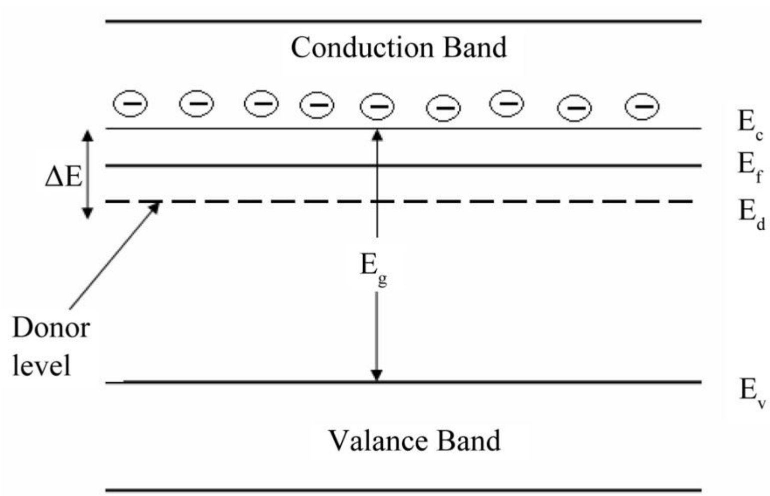
- Disadvantages:
  - Higher energy threshold for production of ionisation than other detectors $\rightarrow$ less ionisation produced for same incident energy $\rightarrow$ lower energy resolution
  - Low Z $\rightarrow$ low stopping power

$\rightarrow$ Limited application in medical imaging

Semiconductor detectors

Ref. 4 – Chapter 10, Ref. 3 – Chapters 3.3 and 6.4

- Crystalline material with energy band structure = valence, conduction bands and ‘forbidden’ energy gap



- Semiconductor detectors –vs– gaseous detectors
  - Advantages:
    1. Only few eV needed to create e–h pair in Si and Ge → for given energy greater ionisation produced than in gas → higher energy resolution
    2. Higher Z than gas → greater stopping power
  - Disadvantages:
    1. More expensive

Signal formation

- Ionising radiation → e–h pair → Charge carriers = e^- in conduction band + h^+ in valence band

$$J = e\, n_i(\mu_e + \mu_h)E$$

J = current density

e = electron charge

n_i = electron / hole concentration

μ_e, μ_h = electron and hole mobility

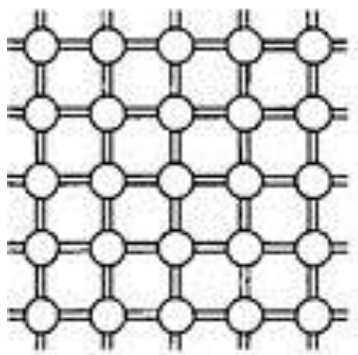
E = electric field

- If material fully depleted = solid-state ionisation chamber → very high detection efficiency for each e–h pair

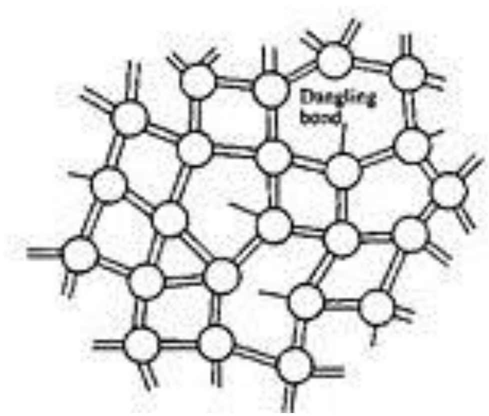
Amorphous semiconductors

http://chemwiki.ucdavis.edu/Textbook_Maps/Inorganic_Chemistry_Textbook_Maps/Map%3A_Inorganic_Chemistry_(Wikibook)/Chapter_10%3A_Electronic_Properties_of_Materials%3A_Superconductors_and_Semiconductors/10.7_Amorphous_semiconductors

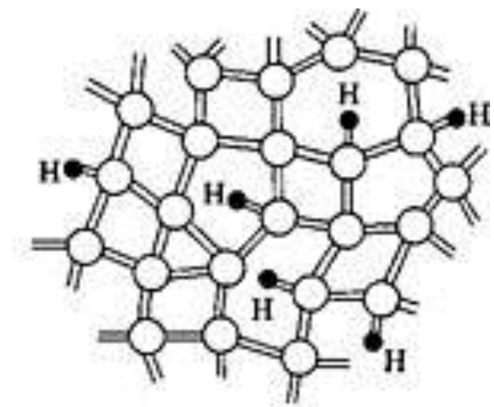
- Disordered or glassy form
- Two most studied: a-Si and a-Se



Crystalline Si



Amorphous Si
a-Si



Amorphous hydrogenated Si
a-Si:H

- Advantage: can be produced in large areas

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Semiconductor materials

	Si	a-Si	a-Se	Ge	CdZnTe
Atomic number	14	14	34	32	48,30,52
Average atomic number	14	14	34	32	49.1
Density (g/cm^3)	2.33	2.30	4.30	5.32	5.78
Bandgap (eV)	1.12	1.80	2.30	0.66	1.572
e^- mobility μ_e (cm^2/Vs)	1400	1	0.005	3900	1000
e^- lifetime τ_e (s)	$>10^{-3}$	7×10^{-9}	10^{-6}	$>10^{-3}$	$3 \div 5 \times 10^{-6}$
h^+ mobility μ_h (cm^2/Vs)	480	0.005	0.14	1900	50÷80
h^+ lifetime τ_h (s)	2×10^{-3}	4×10^{-6}	10^{-6}	10^{-3}	10^{-6}
$\mu_e \tau_e$ (cm^2/V)	>1.4	7.0×10^{-9}	5.0×10^{-9}	>3.9	$3 \div 5 \times 10^{-3}$
$\mu_h \tau_h$ (cm^2/V)	0.96	2.00×10^{-8}	1.40×10^{-7}	1.90	$5 \div 8 \times 10^{-5}$
Ionisation energy (eV/e–h pair)	3.62	4.00	20÷60[1]	7.90	4.64
Dielectric constant	11.7	11.7	6.6	16.0	10.9
Resistivity (Ω cm)	$<10^4$	10^{12}	10^{12}	46	3×10^{10}

[1]Depends on ionising energy

Si planar detectors

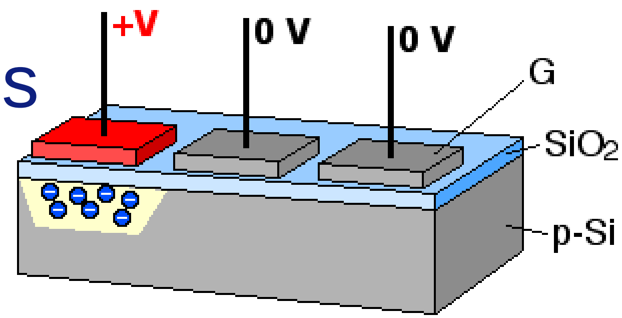
Ref. 3 – Chapter 6.4

- Si material
  - Advantages:
    - Spatial resolution down to few μm for microstrips & pixels
    - Excellent energy resolution = few keV @ room temperature
  - Disadvantages:
    - Low $Z \longrightarrow$ low stopping power
- Two different planar configurations:
  1. Strip detectors = separate read-out strips
  2. Pixel detectors = one of two electrodes segmented in small area cells called pixels

Charge-Coupled Devices (CCDs)

Ref. 2 – Chapter 2.8.3.2

- Integrated circuit = series of electrodes deposited on Si substrate + depletion region below = 2D array of MOS capacitors = 2D array of potential wells
- Ionising radiation → photoelectron created and trapped in wells = charge distribution = image
- Image read out shifting the charge from one well to next until it reaches digitising electronics



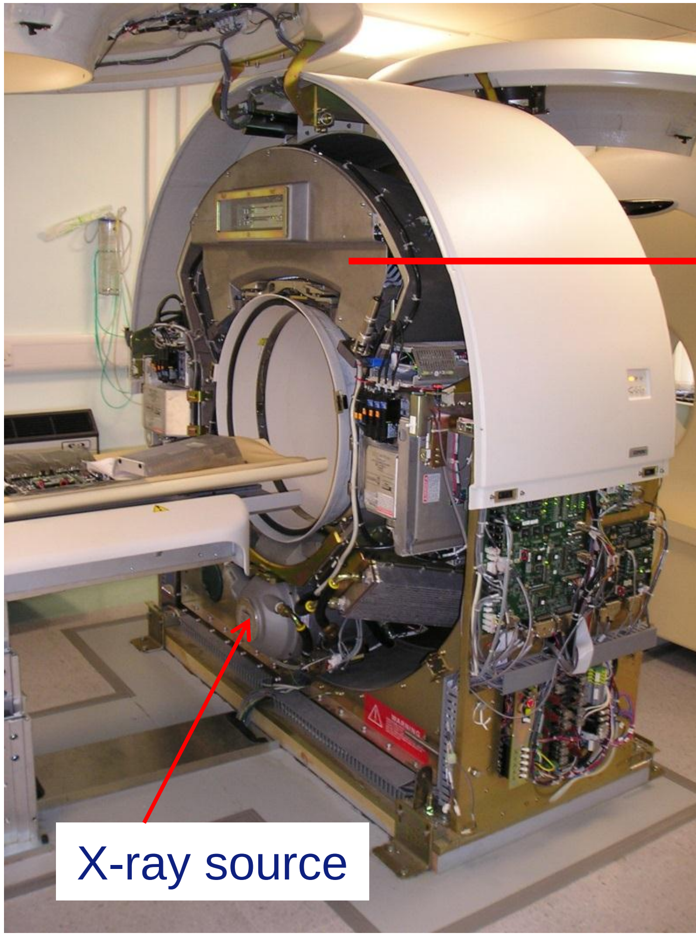
CCDs
Pro and Cons

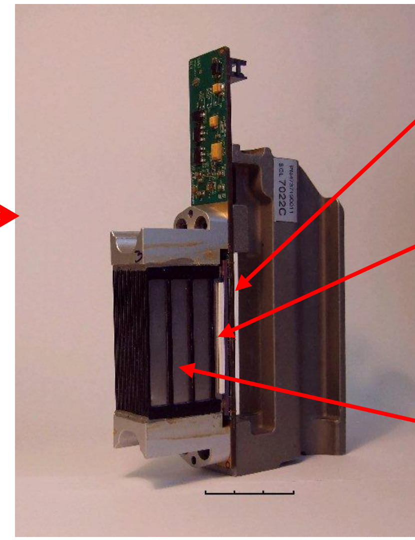
- Advantages:
  - Transfer efficiency very high = approaching 100%
  - Read-out times of 30 frames/s possible with 1000x1000 pixels
- Disadvantages:
  - Small sensitive area limited by engineering to 5 cm. Possible solutions:
    - Separate arrays may be ‘patched’ together but ‘patch’ lines may be visible on images
    - Demagnification optical system $\rightarrow$ better solution
  - Low detection efficiency for photons with $E > 30$ keV and charged particles $\rightarrow$ used coupled with scintillators

Flat panels

- Flat panel detector composed of:
  1. a-Si:H pixel detector: ≅ μm thin a-Si:H sensor coupled to array of thin film transistor (TFT) = high sensitivity to visible light, low sensitivity to X-rays
  2. Coating: $400 \div 500$ μm thick layer of phosphor or scintillator = high sensitivity to X-rays
- Incident X-ray → converted into green light by the coating → green light converted into electric signal by the a-Si:H pixel detector
- Used in general radiography, mammography, fluoroscopy

CT scanners





CT scanner:

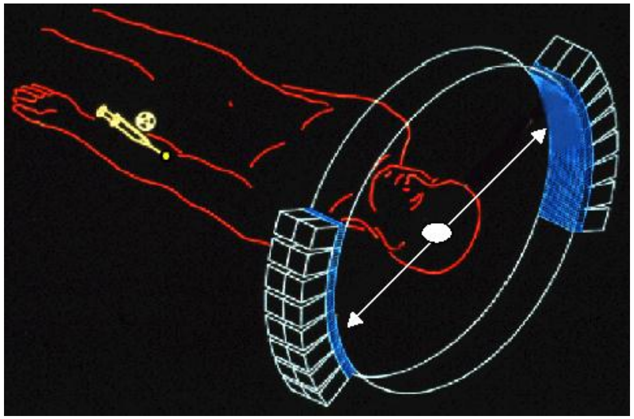
- 120 keV X-rays rotating source
- Diametrically opposite detector unit

Courtesy Mike Partridge (Oxford)

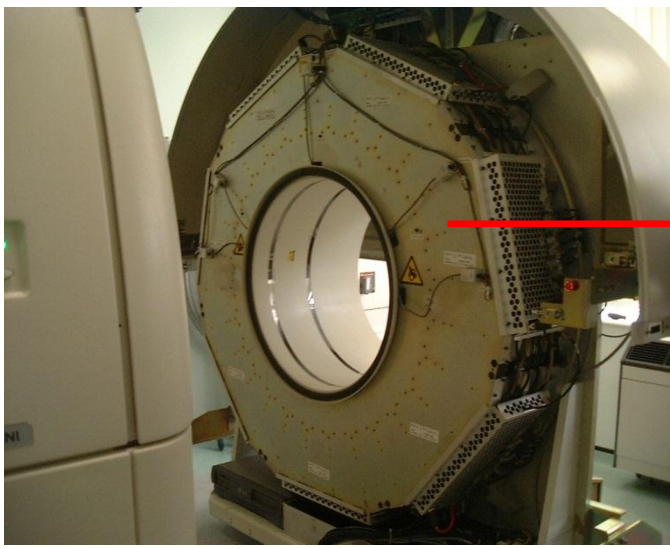
SPECT scanners

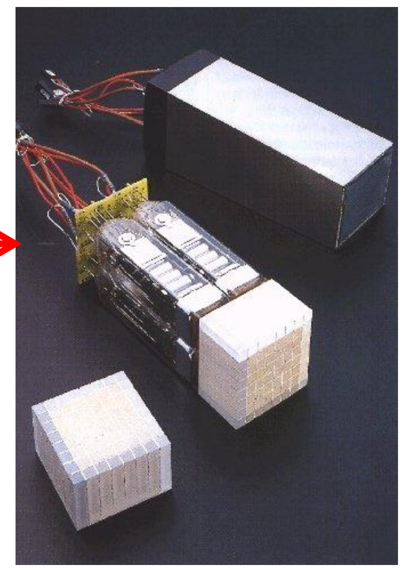
- $^{99}Tc^m$ injected into body $\rightarrow$ one 141 keV γ
- SPECT scanner: two or three rotating detector heads = gamma camera

PET scanners



- β^+ emitter injected into body $\rightarrow$ two back-to-back 511 keV γ from e^+ decay
- PET scanner: ring of detectors

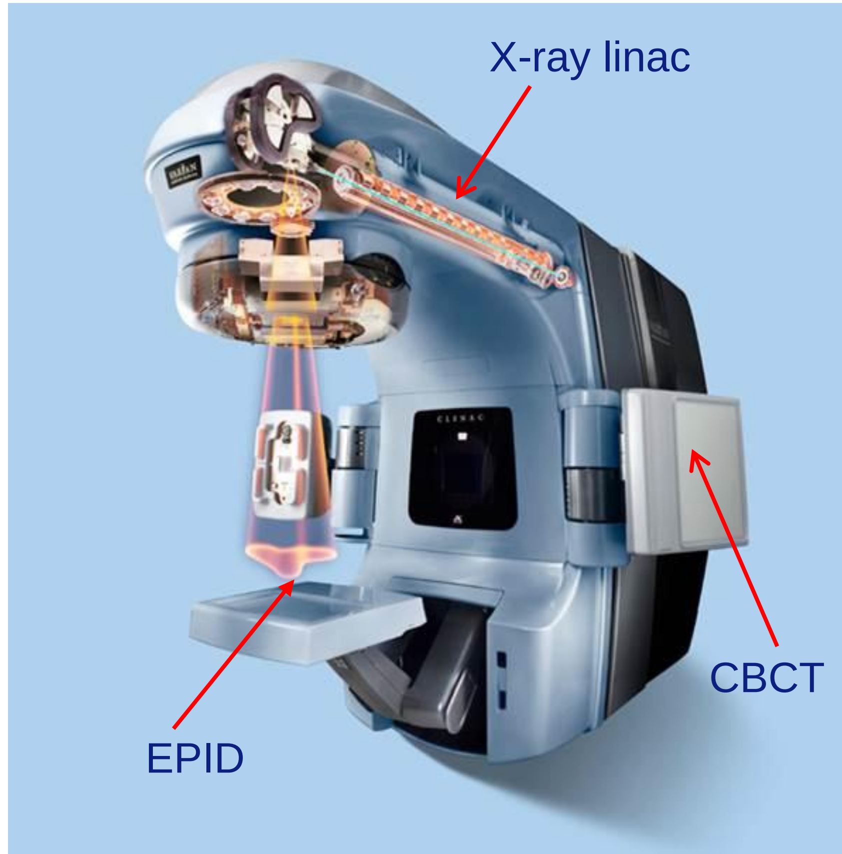




Courtesy Mike Partridge (Oxford)

Imaging systems for X-ray radiotherapy

- Cone Beam CT (CBCT)
  - CT scanner but with X-ray beam diverging (cone)
- Electron Portal Imaging Devices (EPIDs):
  - Used to check patient set-up and field placement
  - Measure therapeutic beam exiting the patient
  - Detectors:
    - Camera based
    - Liquid-ionising-chamber based
    - a-Si flat-panel based



After Bleddyn Jones (Oxford)