Detector technology

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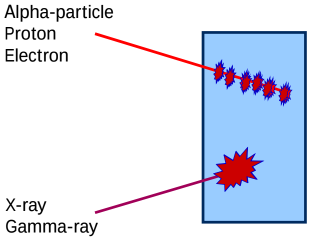
Aim of this talk

> You can know the name of a bird in all the languages of the world, but when you're finished, you'll know absolutely nothing whatever about the bird... So let's look at the bird and see what it's doing - that's what counts. I learned very early the difference between knowing the name of something and knowing something.
>
> Richard Feynman, Nobel Prize Physicist, 1918 - 1988

Principle of a radiation detector

A radiation detector converts the energy of ionizing particles into charge pulses



Alpha-particle
Proton
Electron

X-ray
Gamma-ray

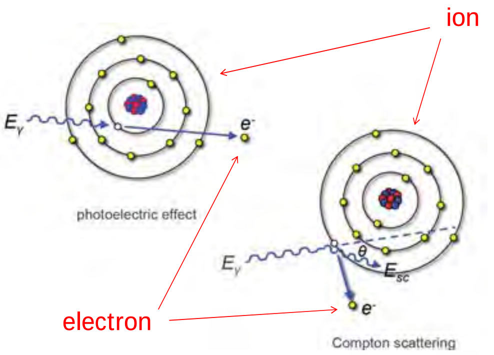
Interaction => Ionization track

Free charge carriers
- gas-filled detectors
- semi-conductors

Conversion to luminescence
- Scintillator + photosensor

=> electric signal

Interactions of gamma photons (gas)



ion

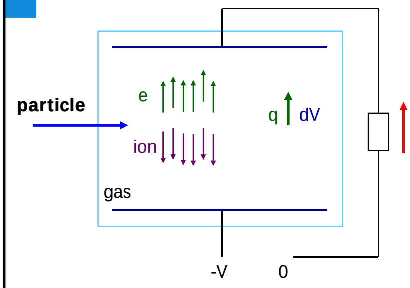
electron

Ionization chamber: gas

charge transport

⇩

signal formation



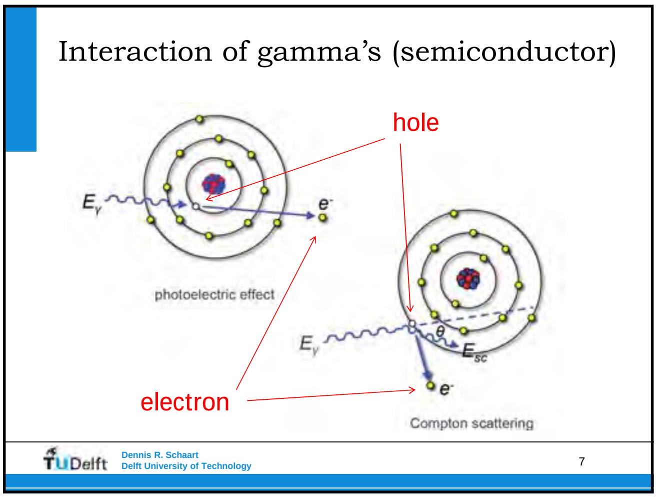
$dq = q dV/V$

$i = dq/dt$

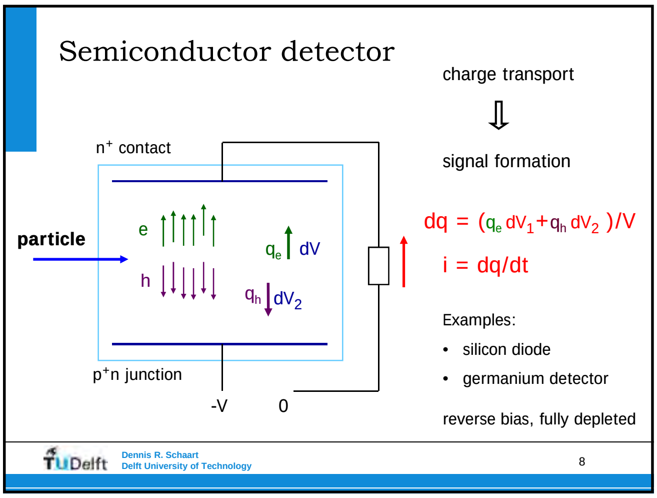
Example:
- free-air dose meter

Gas-filled detectors: examples

Geiger-Muller detector (saturation detection)



Interaction of gamma's (semiconductor)



Semiconductor detector

charge transport

⇩

signal formation

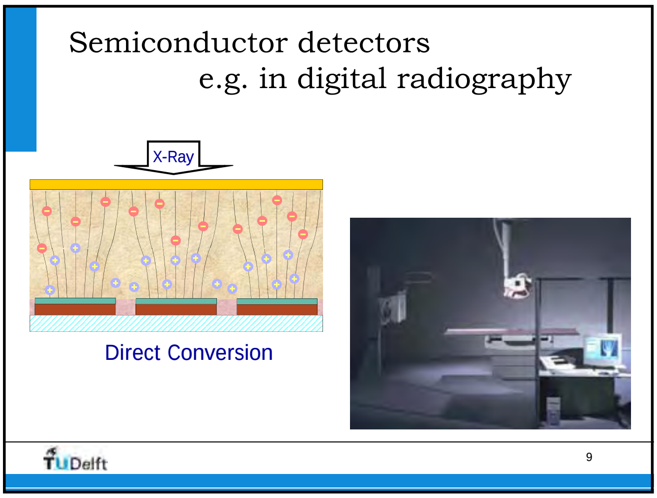
$dq = (q_e\,dV_1 + q_h\,dV_2)/V$

$i = dq/dt$

Examples:

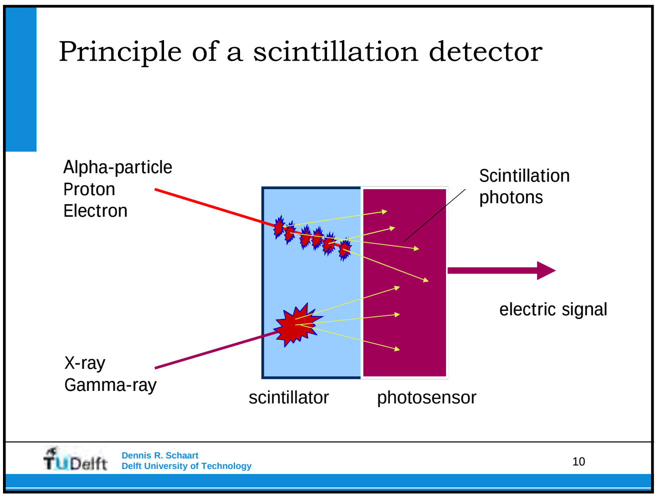
- silicon diode
- germanium detector

reverse bias, fully depleted

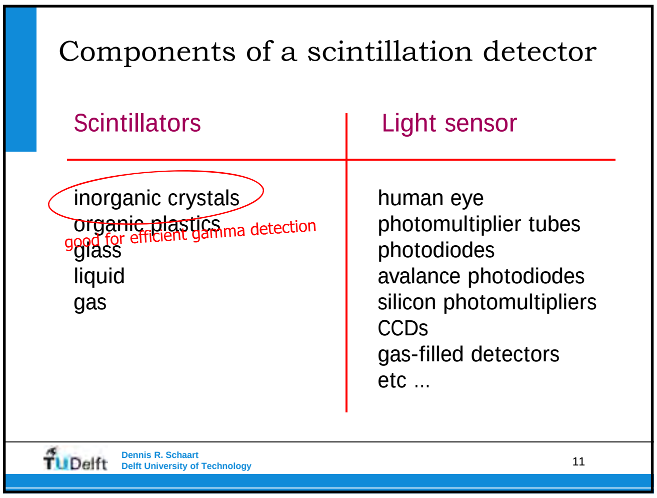


Semiconductor detectors
e.g. in digital radiography

Direct Conversion



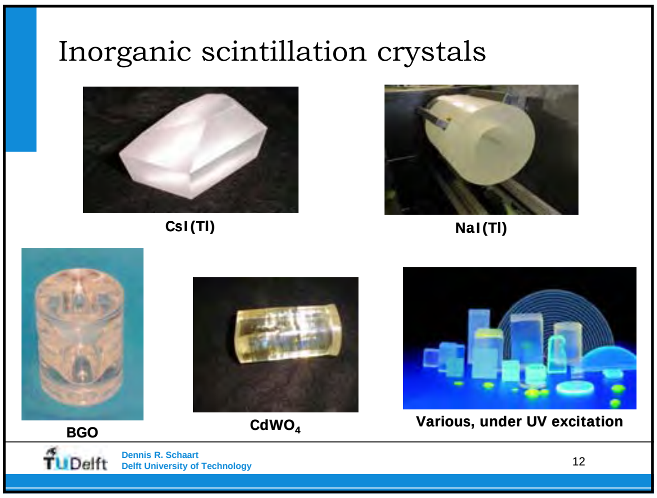
Principle of a scintillation detector



Components of a scintillation detector

Scintillators	Light sensor
inorganic crystals	human eye
organic plastics	photomultiplier tubes
glass	photodiodes
liquid	avalance photodiodes
gas	silicon photomultipliers
	CCDs
	gas-filled detectors
	etc ...

good for efficient gamma detection



Inorganic scintillation crystals

CsI(Tl)

NaI(Tl)

BGO

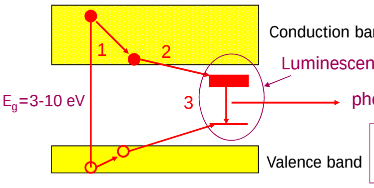
$CdWO_4$

Various, under UV excitation

The scintillation process

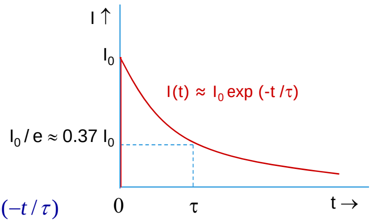
Three phases:

1. The interaction phase + thermalization phase (ps)
2. The charge carrier and energy migration phase (ns-ms)
3. The luminescence phase (ns-μs)



VIS	**400 - 800 nm**	**3.1 - 1.6 eV**
UV	**180 - 400 nm**	**6.9 - 3.1 eV**
VUV	**< 180 nm**	**> 6.9 eV**

Scintillation pulse shape (simple case)



Intensity $I(t) = I_0 \exp(-t/\tau)$

with I_0 given by $\int_0^\infty I(t)dt = E_\gamma Y$

where E_γ is the absorbed gamma energy (in MeV)
and Y the scintillator light yield (in photons/MeV)

Important scintillator parameters

high light output	Y (photons/MeV)
fast scintillation speed	τ (ns)
good energy resolution	R_{FWHM} (%)
high density for γ detection	ρ (g/cm^3)
large size of crystal	10-100-1000 cm^3
low cost per cm^3	
low afterglow (low phosphorescence)	
low background count rate (low intrinsic activity)	

- absence of radioactive isotopes

Relative importance depends on application

Properties of some inorganic scintillators

scintillator	mass density	index of refraction	decay constant	emission (nm)	ph/MeV
NaI (*at 80* K)	3.67	1.75	60	303	76000
NaI(Tl)	3.67	1.75	230	415	38000
CsI(Tl)	4.51	1.75	3340	540	65000
BaF_2	4.89	1.5	630	310	9500
(*valence* e⁻ *to core*)			0.6	220	1400
$Bi_4Ge_3O_{12}$ (BGO)	7.13	2.15	300	480	8200
$PbWO_4$	8.28	2.20	10	470	100
Lu_2SiO_5:Ce (LSO)	7.4	1.8	47	420	25000
$YAlO_3$:Ce (YAP)	5.37	1.95	27	370	18000
$LaCl_3$:Ce	3.7	1.8	35	350	50000
$LaBr_3$:Ce	5.1	2.1	17	380	70000

Properties of some inorganic scintillators

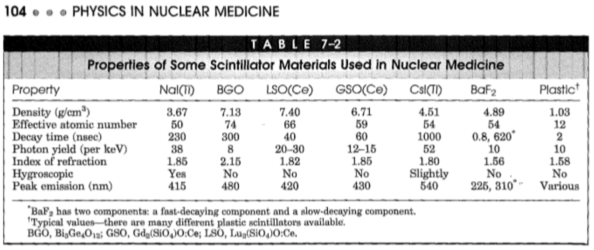


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TABLE 7-2

Properties of Some Scintillator Materials Used in Nuclear Medicine

Property	NaI(Tl)	BGO	LSO(Ce)	GSO(Ce)	CsI(Tl)	BaF_2	Plastic[†]
Density (g/cm^3)	3.67	7.13	7.40	6.71	4.51	4.89	1.03
Effective atomic number	50	74	66	59	54	54	12
Decay time (nsec)	230	300	40	60	1000	0.8, 620[*]	2
Photon yield (per keV)	38	8	20–30	12–15	52	10	10
Index of refraction	1.85	2.15	1.82	1.85	1.80	1.56	1.58
Hygroscopic	Yes	No	No	No	Slightly	No	No
Peak emission (nm)	415	480	420	430	540	225, 310[*]	Various

[*] BaF_2 has two components: a fast-decaying component and a slow-decaying component.
[†] Typical values—there are many different plastic scintillators available.
BGO, $Bi_3Ge_4O_{12}$; GSO, $Gd_2(SiO_4)O$:Ce; LSO, $Lu_2(SiO_4)O$:Ce.

S.R. Cherry, J.A. Sorenson, M.A. Phelps, Physics of Nuclear Medicine, 3rd ed., 2003

Photomultiplier tubes

The main PMT elements

- photocathode
  - photon → photoelectron
  - quantum efficiency (e.g. ~25%)
- electron-optics
  - focusing of photoelectron on the first dynode
  - preservation of time information
- multiplication stage
  - 6 to 12 dynodes
- charge pulse at anode output

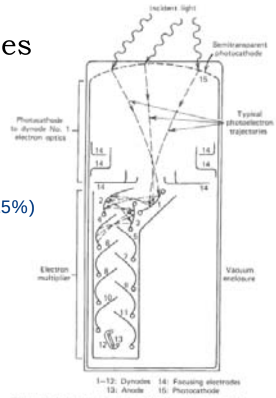
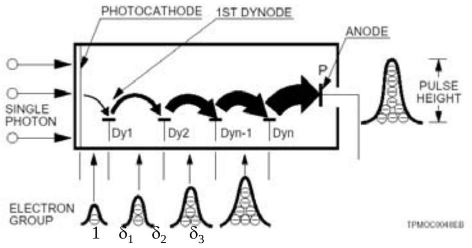


Figure 9.1 Basic elements of a PM tube. (From Ref. 1.)

Conversion and multiplication



Quantum efficiency of photocathode 0.01 – 0.4 photoelectrons/photon
Overall electron gain is sensitive to applied voltage (typically 1 kV – 2.5 kV)
Secondary emission factor of dynodes δ typically 4-8
Typical gain = 10^6-10^7 (number of dynodes N = 8-12)

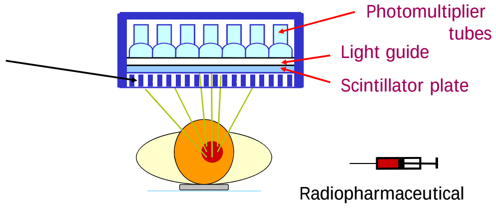
Scintillation detectors in nuclear medicine

- Gamma camera (planar scintigraphy)
- SPECT scanner (single photon emission computed tomography)
- PET scanner (positron emission tomography)
  - Time-of-Flight PET (TOF-PET)

Scintillation detectors in nuclear medicine

Simplest case:
planar scintigraphy

2D position sensitive
detector: gamma camera



Photomultiplier tubes

Light guide

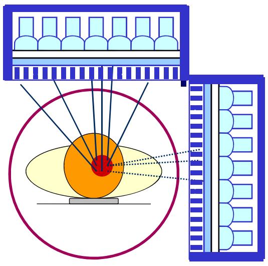
Scintillator plate

Collimator

Radiopharmaceutical

Single Photon Emission CT (SPECT)

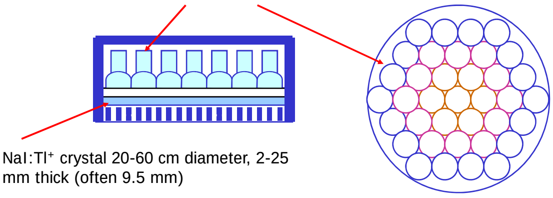
rotation



Gamma camera

NaI:Tl crystal of gamma camera

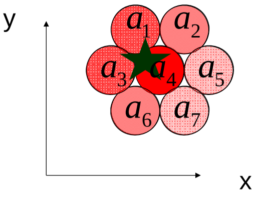
7 to 61 photomultipliers



NaI:Tl⁺ crystal 20-60 cm diameter, 2-25 mm thick (often 9.5 mm)

Position Estimation (Anger logic)

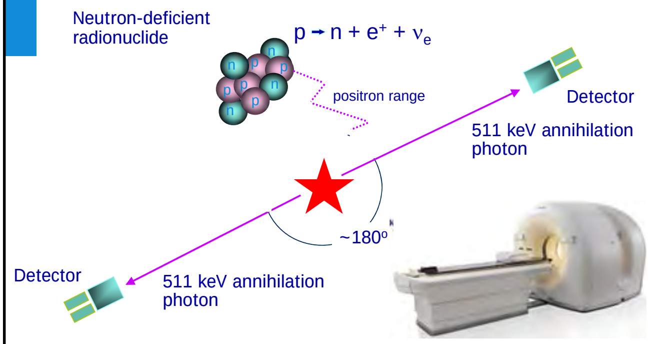
$a_1 = 0.8, \quad a_2 = 0.5$
$a_3 = 0.8, \quad a_4 = 1$
$a_5 = 0.3, \quad a_6 = 0.5$
$a_7 = 0.1$



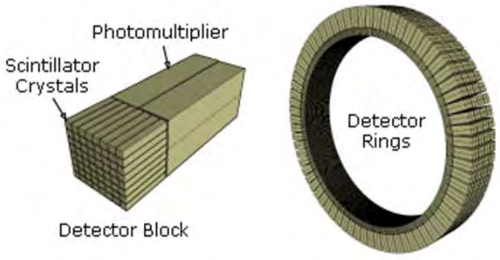
Energy: $E = \sum_{k=1}^{7} a_k$

Position: $X = \frac{1}{E} \sum_{k=1}^{7} a_k x_k \qquad Y = \frac{1}{E} \sum_{k=1}^{7} a_k y_k$

Positron emission tomography (PET)



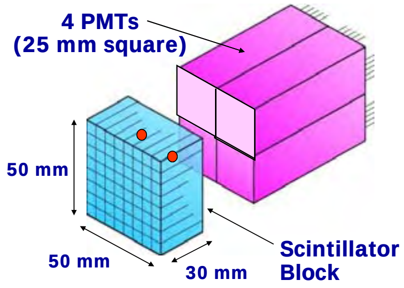
PET detectors: classic "block" detector



- Several block detectors are assembled into a ring
- A scanner may consist of several detector rings

PET detectors: classic "block" detector

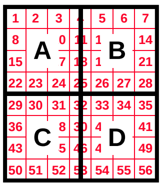
- Saw cuts direct light toward PMTs.
- Depth of cut determines light spread at PMTs.
- Crystal of interaction found with Anger logic (i.e. PMT light ratio).

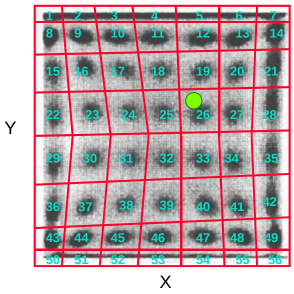


Courtesy of Bill Moses, LBNL

PET detectors: Anger logic

- Identify crystal of interaction using lookup table
- Position given by crystal ID





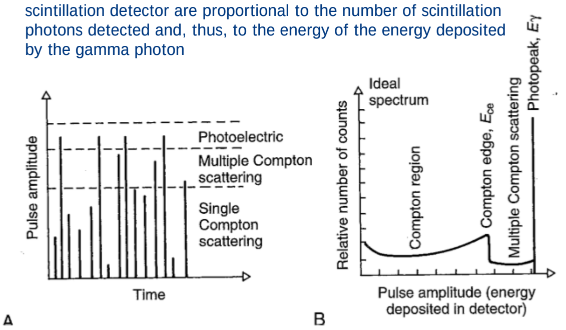
Energy: $E = A + B + C + D$

Position: $Y = (A + B) / E$
$X = (B + D) / E$

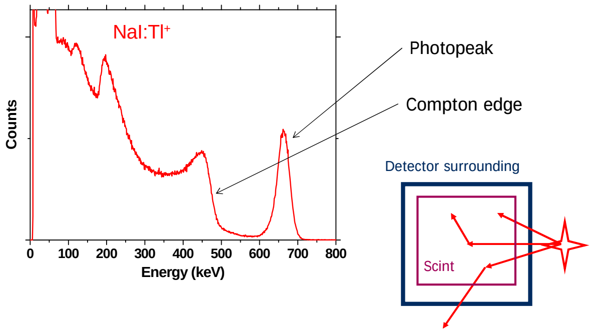
Courtesy of Bill Moses, LBNL

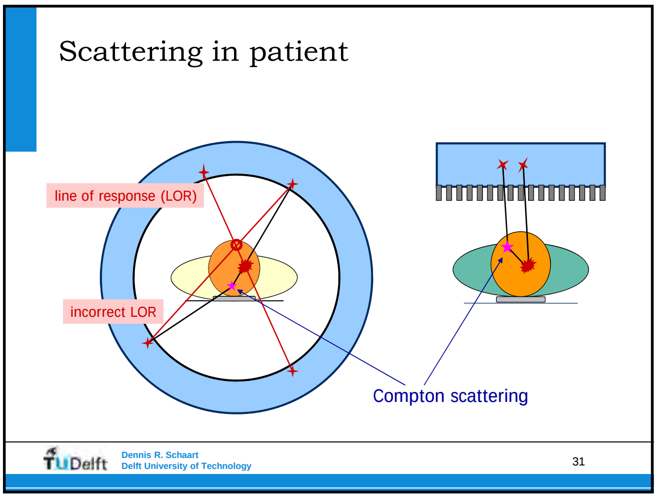
Energy: pulse height spectrum

The height (amplitude) of the charge pulses produced by a scintillation detector are proportional to the number of scintillation photons detected and, thus, to the energy of the energy deposited by the gamma photon

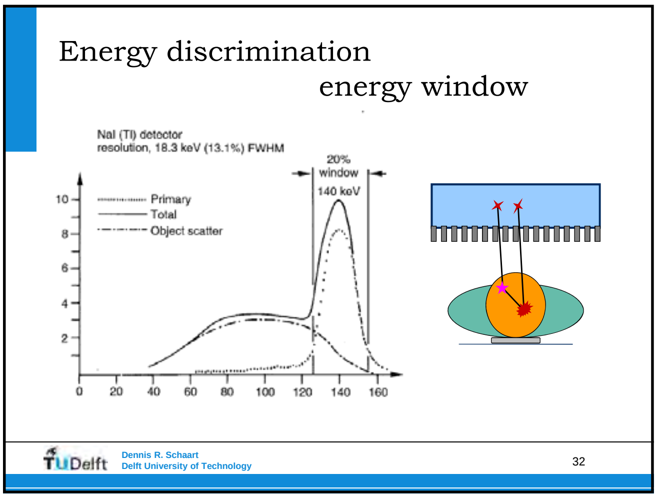


Example: NaI:Tl pulse height spectrum



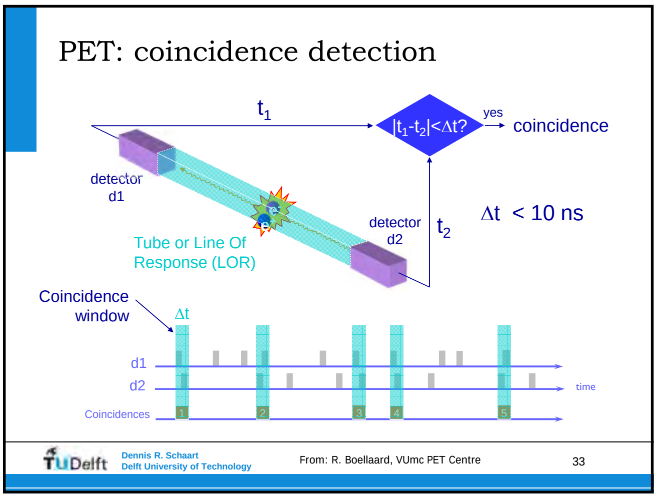


Scattering in patient



Energy discrimination

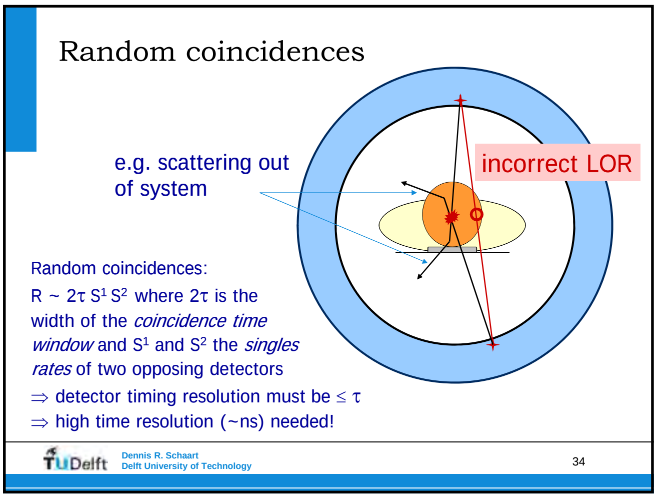
energy window



PET: coincidence detection

$\Delta t < 10$ ns

From: R. Boellaard, VUmc PET Centre



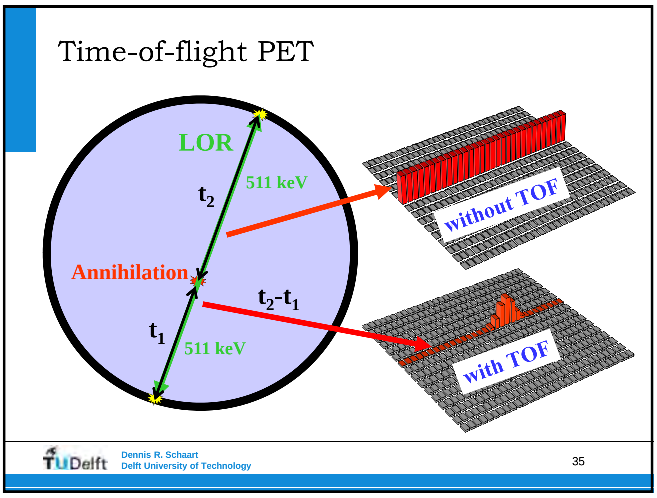
Random coincidences

Random coincidences:

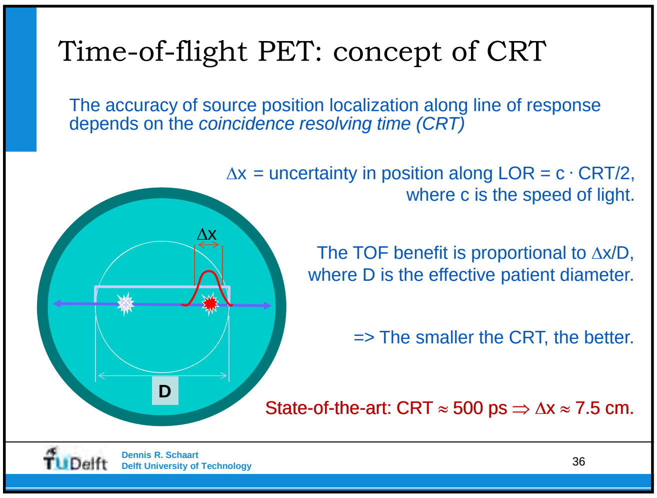
$R \sim 2\tau S^1 S^2$ where 2τ is the width of the *coincidence time window* and S^1 and S^2 the *singles rates* of two opposing detectors

$\Rightarrow$ detector timing resolution must be $\leq \tau$

$\Rightarrow$ high time resolution (~ns) needed!



Time-of-flight PET



Time-of-flight PET: concept of CRT

The accuracy of source position localization along line of response depends on the *coincidence resolving time (CRT)*

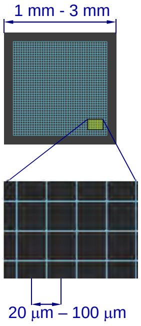
Δx = uncertainty in position along LOR = $c \cdot$ CRT/2, where c is the speed of light.

The TOF benefit is proportional to $\Delta x/D$, where D is the effective patient diameter.

=> The smaller the CRT, the better.

State-of-the-art: CRT ≈ 500 ps ⇒ $\Delta x \approx 7.5$ cm.

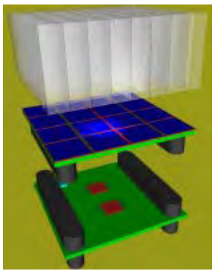
Silicon Photomultiplier (SiPM)



- Array of many single-photon avalanche diodes (microcells) connected in parallel

- Increasingly interesting as replacement for PMTs:
  - high gain (~10^6)
  - high PDE
  - compact and rugged
  - transparent to γ-photons
  - fast response (ns)
  - insensitive to magnetic fields

SiPM-array based PET detectors



For example:

- crystal matrix composed of e.g. 4 mm x 4 mm x 20 mm crystals
- each crystal coupled 1-to-1 to an individual SiPM

=> high spatial resolution
=> high energy resolution
=> excellent timing

Image courtesy of Philips

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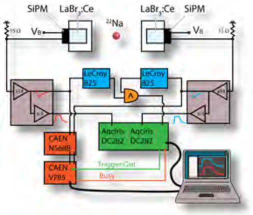
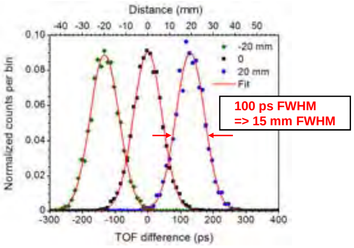
100 ps barrier broken using SiPMs

Made possible by the combination of:

- Small $LaBr_3$:Ce(5%) crystals (3 mm x 3 mm x 5 mm)
- Silicon Photomultipliers (Hamamatsu MPPC-S10362-33-050C)
- Digital Signal Processing (DSP)



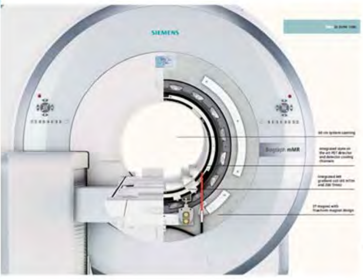


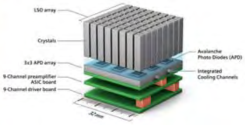
100 ps FWHM
=> 15 mm FWHM

D.R. Schaart et al, Phys Med Biol 55, N179-N189, 2010

Multimodality: PET + MRI

Now: avalanche photodiodes (APDs)
Next generation systems: SiPMs !!!





Images: Siemens