SiPM
AX-PET

Teilchenphysik Seminar

Silicon Photomultipliers

Principle, Performance and Some Applications

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Photodetection

The “classical” application domains of photodetectors in particle physics

- ❑ Calorimetry
  - ○ Readout of organic and inorganic scintillators, lead glass, scint. or quartz fibres → Blue/VIS, usually 10s – 10000s of photons
- ❑ Particle Identification
  - ○ Detection of Cherenkov light → UV/blue → single photons
  - ○ Time Of Flight → Usually readout of organic scintillators (not competitive at high momenta)
  - ○ Transition radiation (X-rays, not covered in this talk)
- ❑ Tracking
  - ○ Readout of scintillating fibres, blue-green, few photons

❖ 158,000 photodetectors in the LHC experiments
❖ 4,45 M readout channels
❖ 35.4 m^2 photosensitive area

C.J., Photodetection in the LHC experiments, NIM A *695 (2012) 13-22*

… and beyond particle physics

- ❑ Medical imaging
  - ○ Nuclear Medicine: PET & SPECT (γ-camera) → Readout of inorganic scintillators
- ❑ Analytical, industrial, …

A novel type of photodetector has emerged over the last ~15 years (too late for LHC!)

- by now reached a reasonable level of maturity
- a number of attractive features, but also some serious limitations.
- It has many names like Silicon PM (SiPM), Geiger APD (G-APD), Metal Resistive layer Semiconductor (MRS-APD), Multi Photon Pixel Counter (MPPC), SiPM, SPM,…

SiPM is by now the most used name (I don't like it).

Outline

- Requirements on photodetectors
- Principle of operation of the Silicon Photomultiplier
- AX-PET – Demonstration of an axial PET scanner
- The Digital SiPM
- Extension of AX-PET by TOF
- SiPM R&D for a SciFi tracker for LHCb and Calorimetry for ILC

Requirements on photodetectors

SiPM AX-PET

"Typical" requirements on a photodetector

- High sensitivity $S(\lambda) = \frac{I[A]}{P[W,\lambda]}$ or $QE(\lambda) = \frac{N_{pe}}{N_\gamma(\lambda)}$

$$S[mA/W] \approx \frac{QE[\%] \cdot \lambda[nm] \cdot e}{hc} = \frac{QE[\%] \cdot \lambda[nm]}{124}$$

- What really counts: Photon detection Efficiency $PDE = QE \cdot \varepsilon_{coll} \cdot \varepsilon_{aval} \cdot \varepsilon_{fill} \cdot \varepsilon \ldots$

- High gain $M = \frac{N_e}{N_{pe}}$ M > 10^5 → we have a chance to see single photons

- Small signal fluctuations, low Excess Noise Factor F $ENF = \frac{\sigma_{out}^2}{\sigma_{in}^2}$ $ENF = 1 + \frac{\sigma_M^2}{M^2}$

  ENF impacts energy resolution $\frac{\sigma_E}{E} = \sqrt{\frac{ENF}{N_{pe}}}$

  general definition (M = 1) (M ≠ 1)

- Low dark noise / current

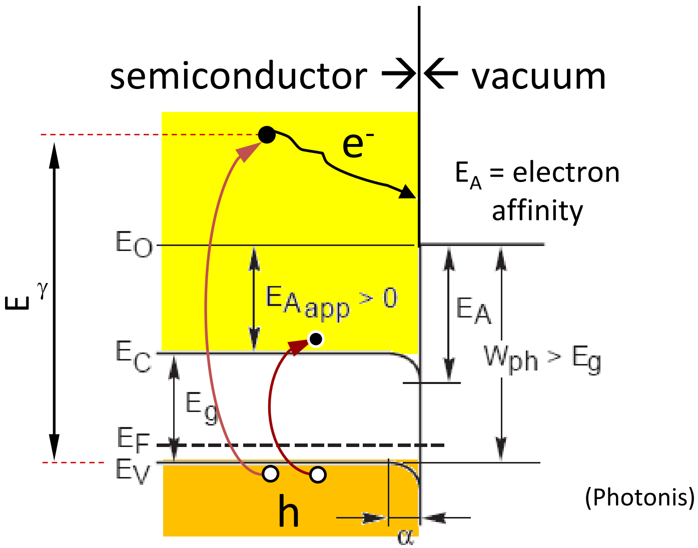
- Fast timing properties (rise time, fall time, dead time)

- Large dynamic range with good linearity

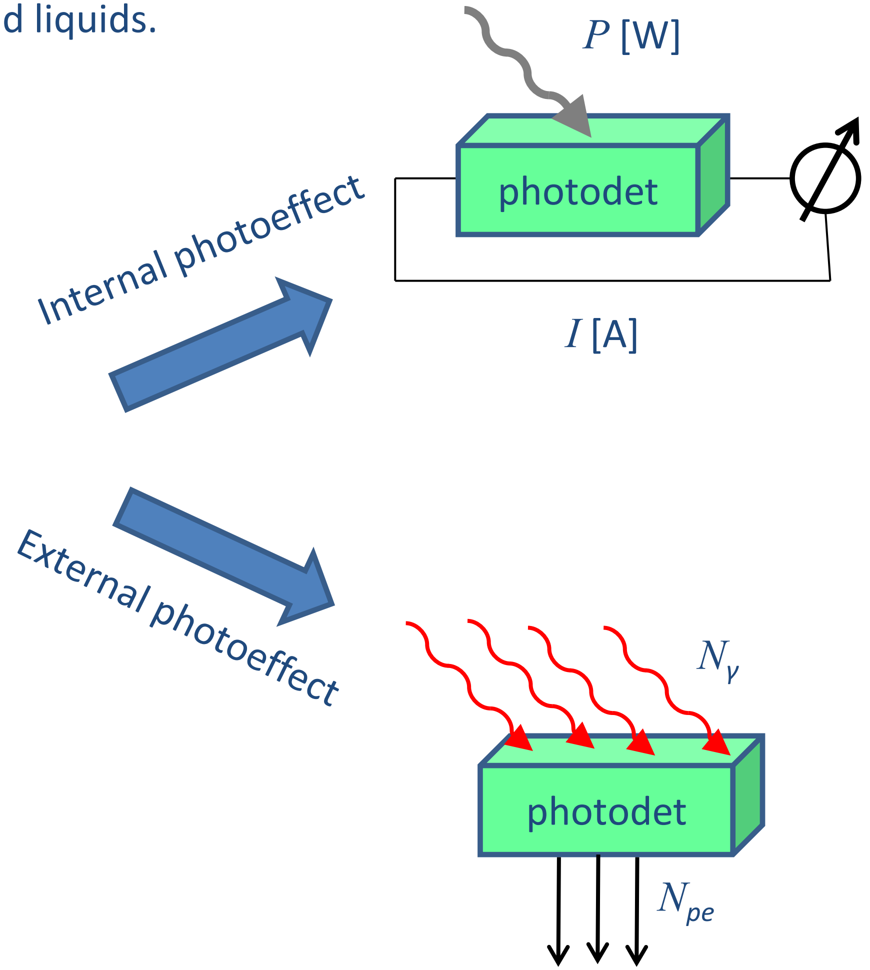
- For certain applications: Radiation hardness, magnetic field compatibility, compactness, cost

Basics of photodetection

Many photosensitive materials are semiconductors, but photoeffect can also be observed from gases and liquids.



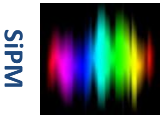
(Photonis)



Internal photoeffect: $E_\gamma > E_g$

External photoeffect: $E_\gamma > E_g + E_A$

Family tree of photo-detectors



Photodetectors

Gas
External photoeffect

TMAE, TEA, CsI + [MWPC, GEM, …]

"Vacuum"
External photoeffect

Avalanche gain Process

Dynodes → PMT

Continuous dynode
→Channeltron, MCP

Multi-Anode devices

Other gain process = Hybrid tubes

Silicon anode	Lumines-cent anodes
HPD	SMART/Quasar
HAPD	X-HPD
(G-APD-HPD)	

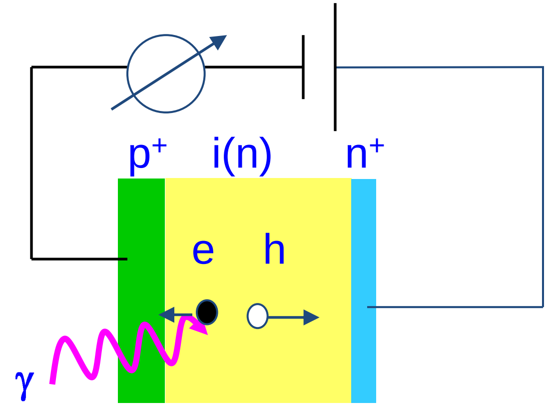
Doesn't exist as 'real' device
Proposed by G. Barbarino et al., NIM A 594 (2008) 326–331
Proof of principle by CJ et al., NIM A 621 (2010) 171-176

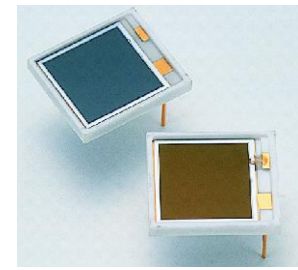
Solid state
Internal photoeffect

PIN-diode
APD
SiPM
Digital SiPM
CMOS imager
CCD

(Si) – Photodiodes (PIN diode)

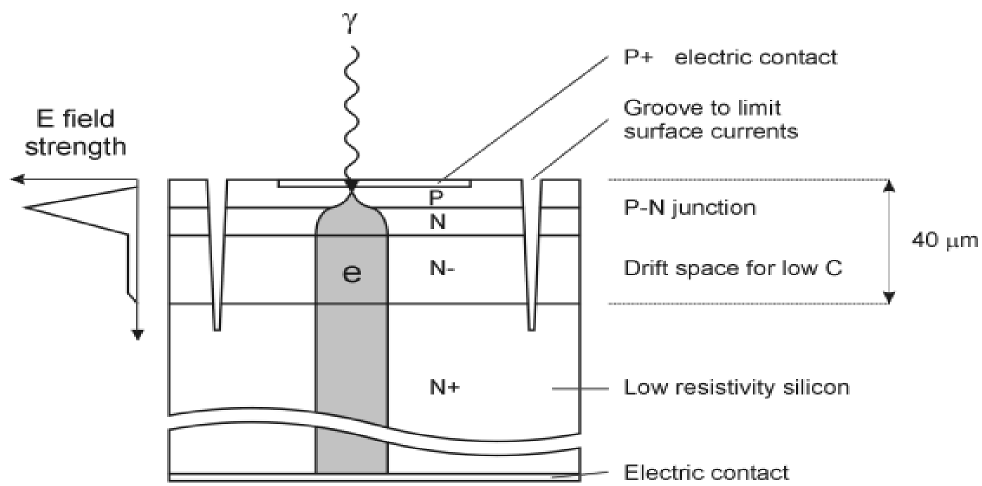
- P(I)N type
- p layer very thin (<1 μm), as visible light is rapidly absorbed by silicon
- High QE (80% @ $\lambda \approx$ 700nm)
- Gain = 1





Avalanche photodiode (APD)

- High reverse bias voltage: typ. few 100 V
- Special doping profile → high internal field ($>10^5$ V/cm) → avalanche multiplication
- Avalanche stops due to statistical fluctuations.
- Gain: typ. O(100)
- Rel. high gain fluctuations (excess noise from the avalanche). CMS ECAL APD: ENF = 2 @G=50.
- Very high sensitivity on temp. and bias voltage ΔG = 3.1%/V and -2.4 %/K



Hamamatsu S8148.
(140.000 pieces used in CMS barrel ECAL).

PIN → APD → Geiger mode Avalanche Photodiode (GM-APD)

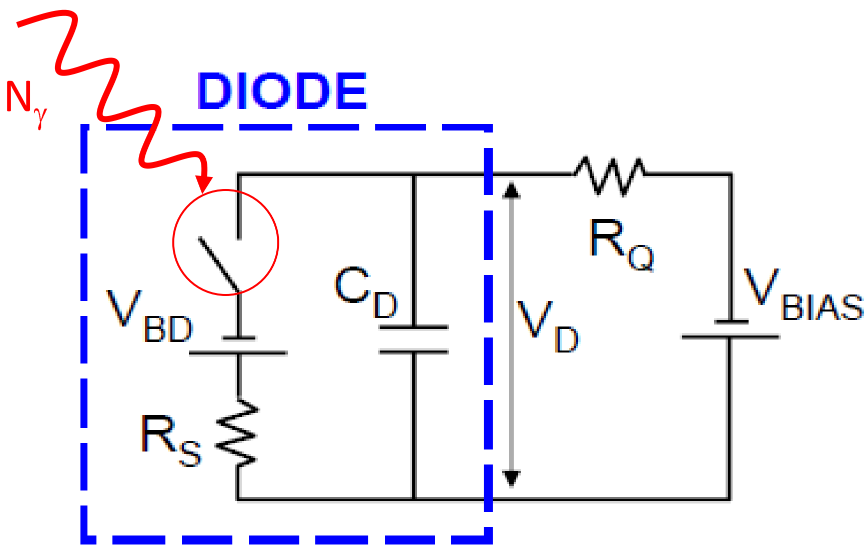
SiPM

AX-PET

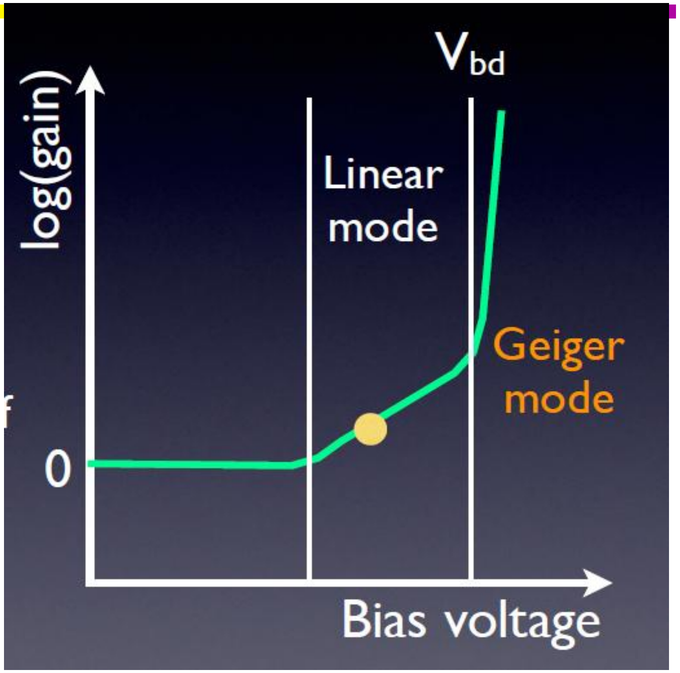
How to obtain higher gain (= single photon detection) without suffering from excessive noise ?

Operate APD cell in Geiger mode (= full discharge), however with (passive/active) **quenching**.

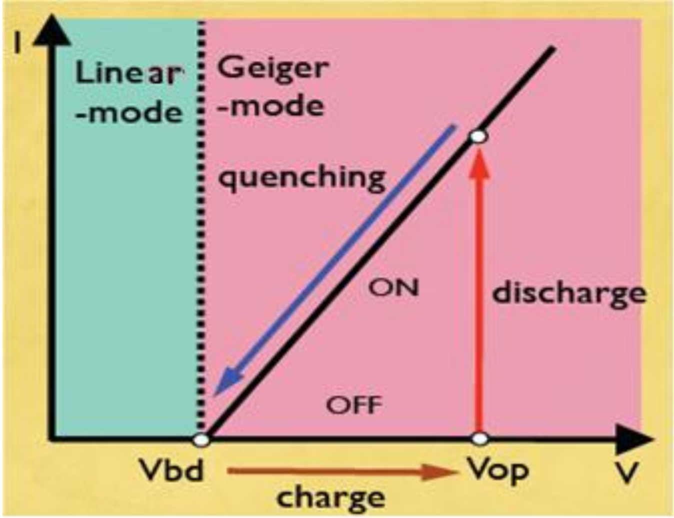
Photon conversion + avalanche short circuit the diode. A single photon (or anything else) is sufficient!



J. Haba, RICH2007



J. Haba, RICH2007



J. Haba, RICH2007

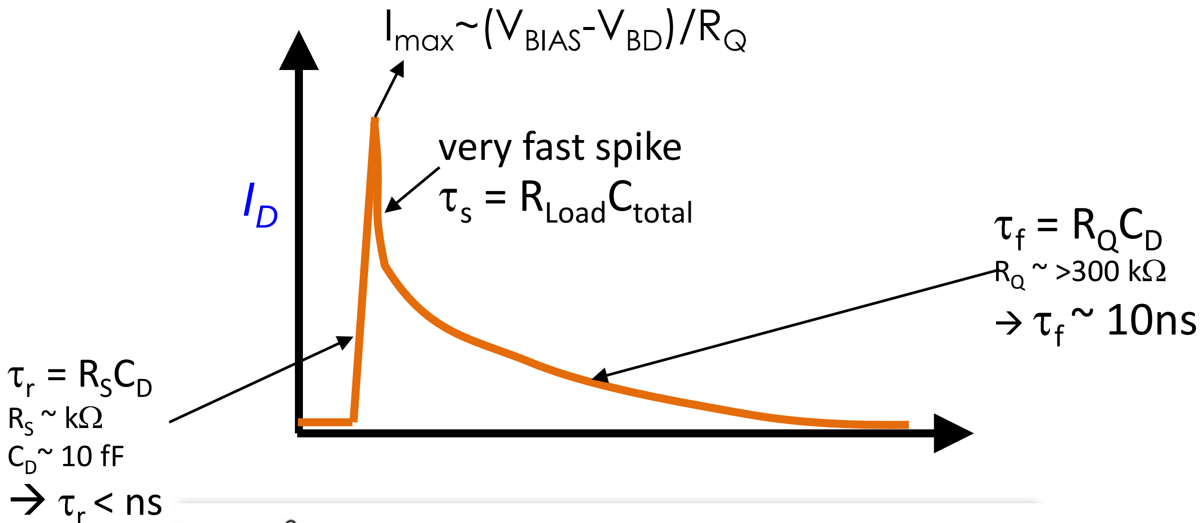
A single-cell GM-APD is just a **binary** device (=switch).
Info on N_γ is lost by the Geiger avalanche.
It will become more interesting when we combine many cells in one device …

Signal characteristics and Gain of a GM-APD

SiPM

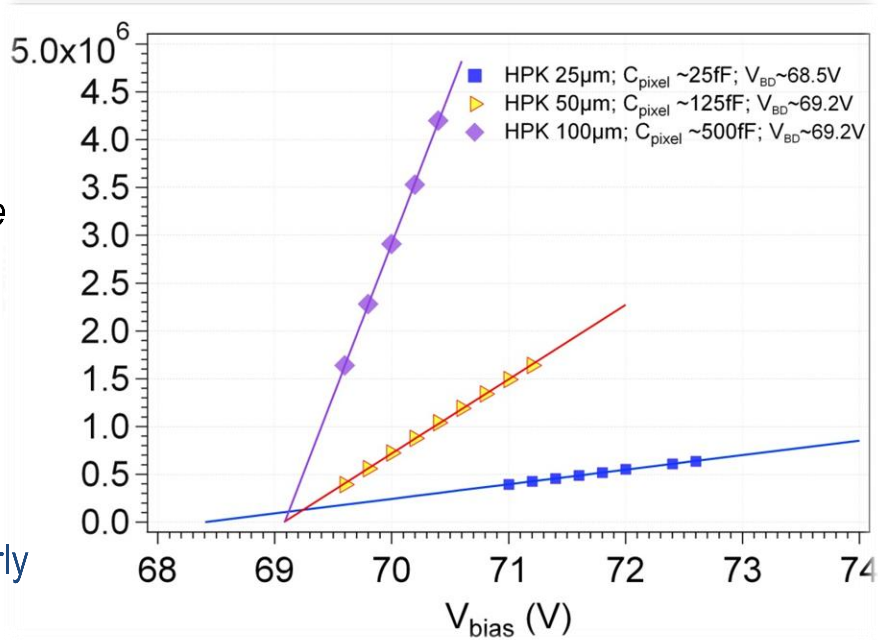
AX-PET

The avalanche formation is intrinsically very fast, because confined to a small space.



$$Gain = Q / e$$
$$= I_{max} \cdot \tau \ / \ e$$
$$= \underbrace{(V_{BIAS}-V_{BD})}_{\Delta V \text{ (overvoltage)}} C_D / e$$

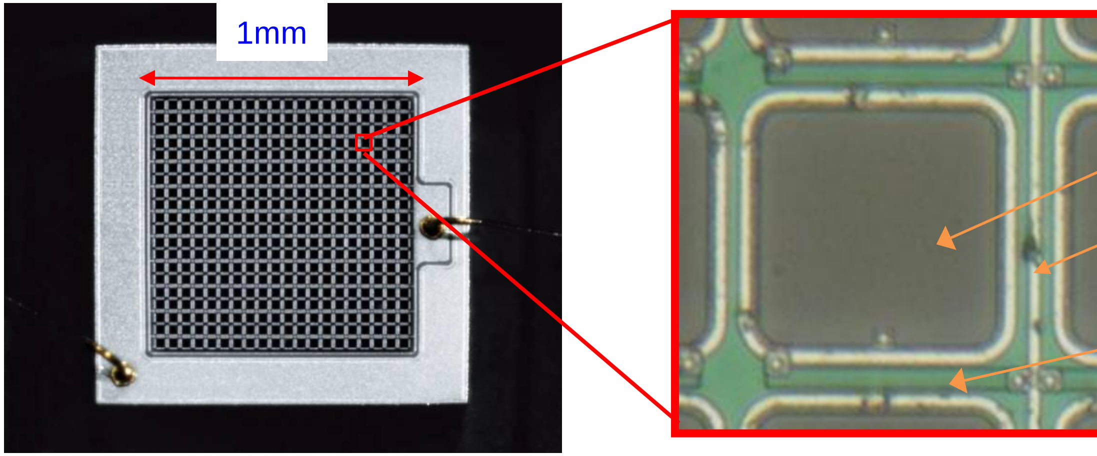
- G ~ 10^5 - 10^6 at rel. low bias voltage (<100 V)
- dG/dT and dG/dV similarly critical as for APD.



- The slope of the linear fit of $G(V_{bias})$ is the pixel capacitance
- C_{pixel} and hence G increase with the pixel geometrical dimensions and the depletion depth.
- $C_{pixel} \sim \varepsilon_0 \varepsilon_r S/d$
- C_{pixel} = O(10-100 fF)

N. Dinu & al, NIM A 610 (2009) 423–426

Multi pixel GM-APD, called SiPM, G-APD, MPPC, …



100 – several 1000 pix / mm^2

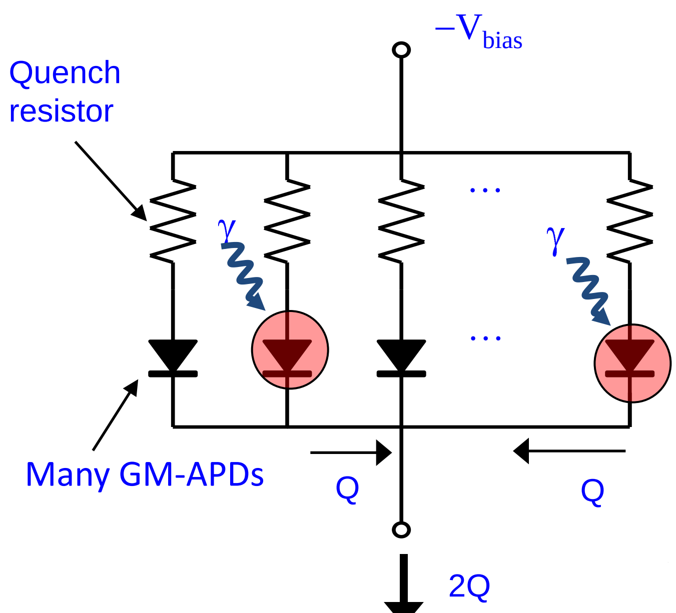
GM-APD

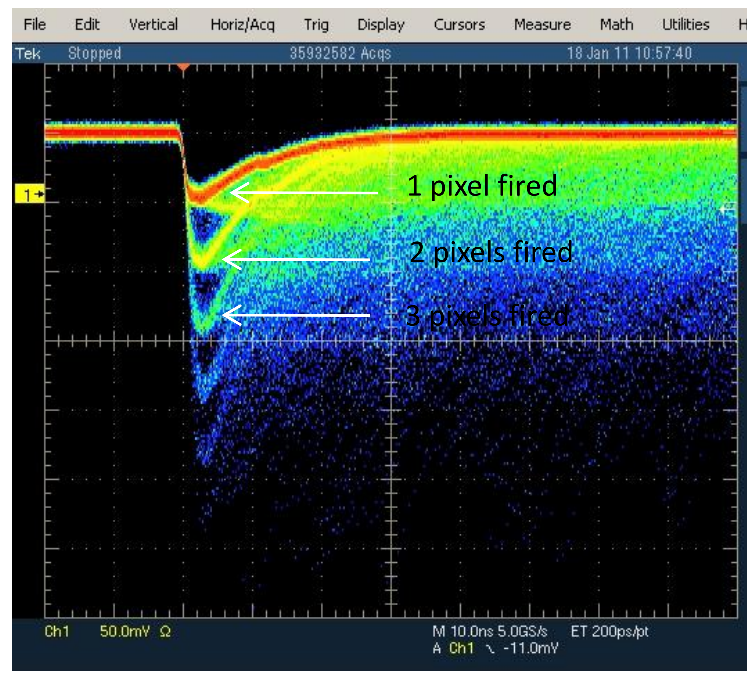
Bias bus

Quench resistor

Only part of surface is photosensitive!

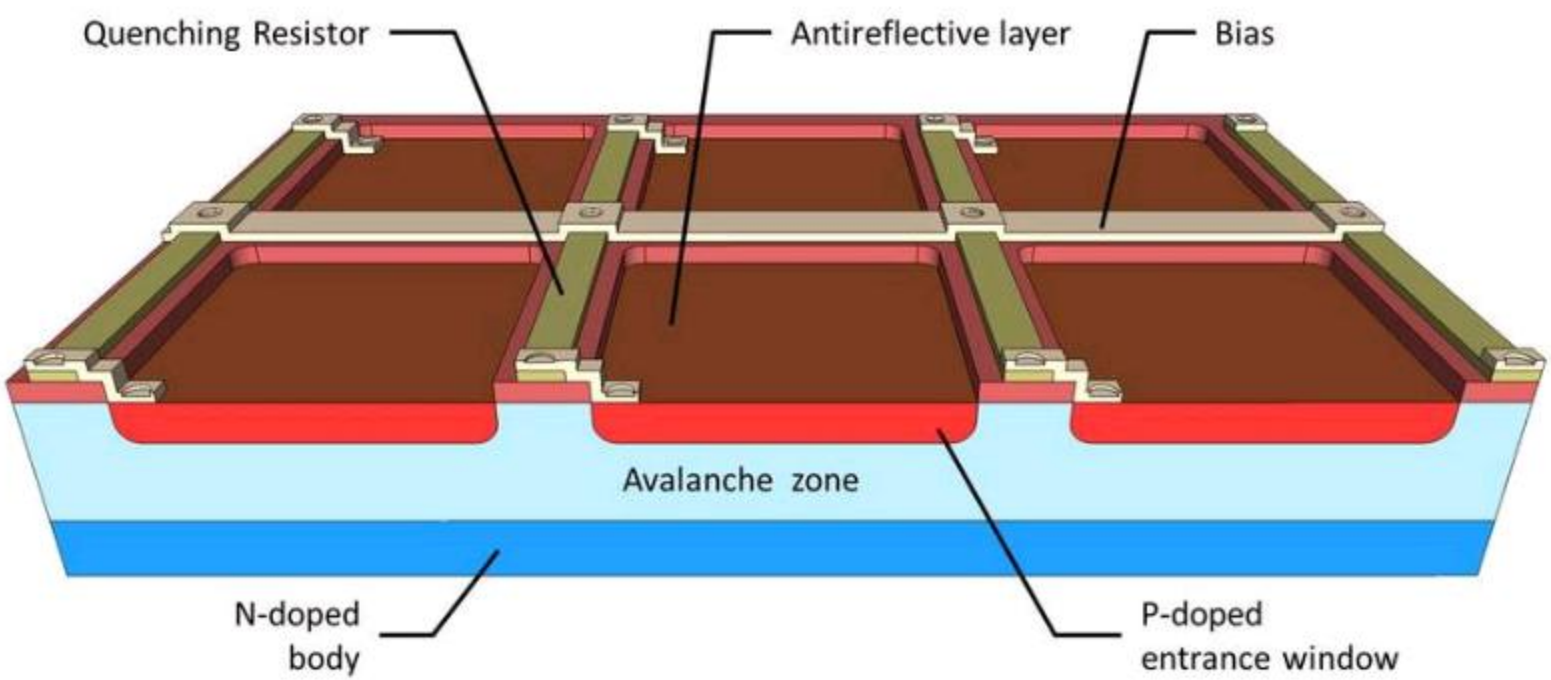
Sizes up to 6×6 mm^2 now standard.





The operation of many binary devices in parallel leads to a quasi-analog detector with very nice properties (photon counting).

Schematic representation of a blue sensitive pn structrue



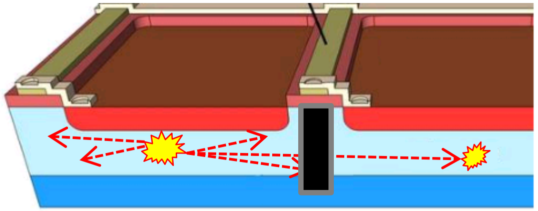
Source: http://www.ketek.net/products/sipm-technology/microcell-construction/

Dark counts due to …

SiPM

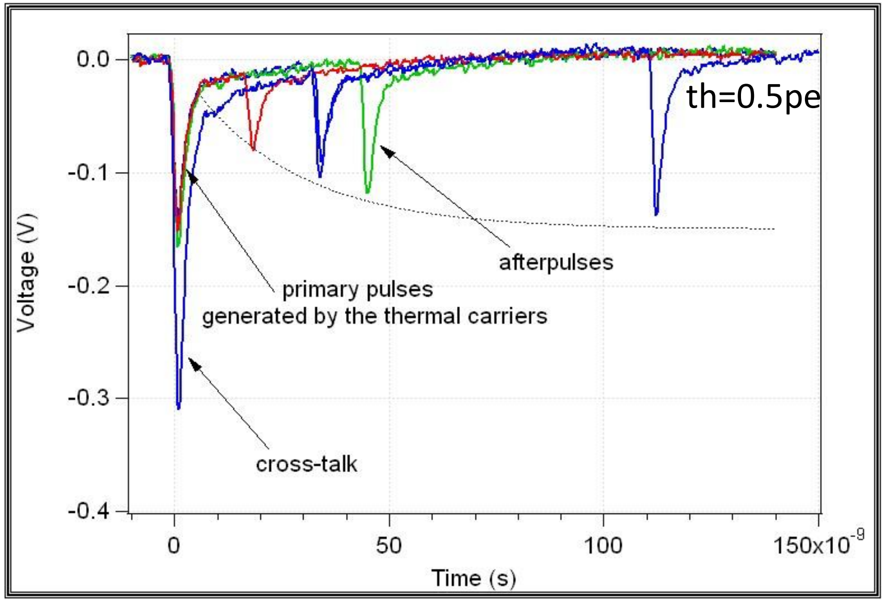
AX-PET

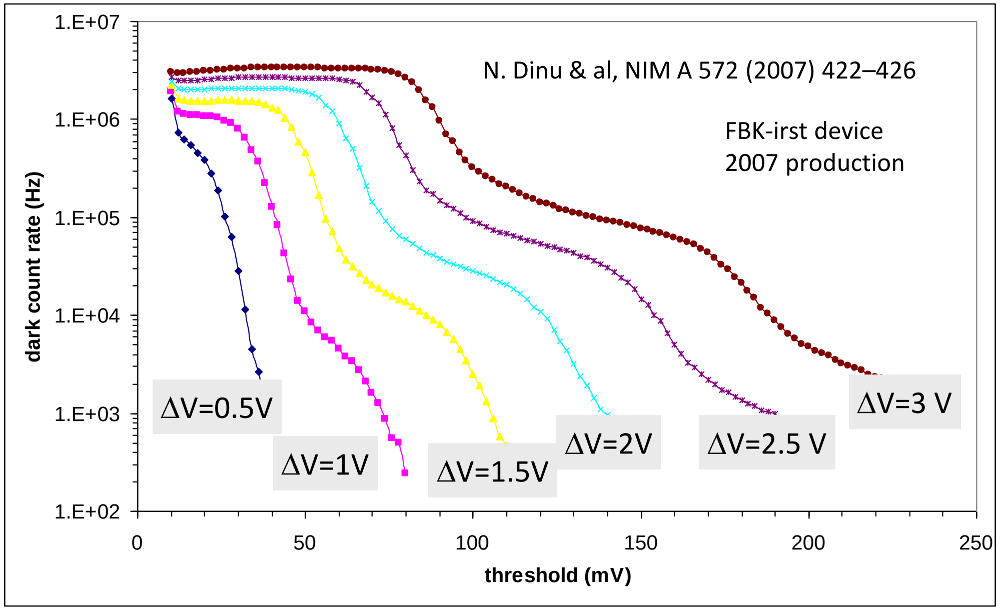
- **<u>Thermal/tunneling</u>** : thermal/ tunneling carrier generation in the bulk or in the surface depleted region around the junction
- **<u>After-pulses :</u>** carriers trapped during the avalanche discharging and then released triggering a new avalanche during a period of several 100 ns after the breakdown
- **<u>Optical cross-talk:</u>** 10^5 carriers in an avalanche plasma emit on average 3 photons with an energy higher than 1.14 eV (A. Lacaita et al. IEEE TED 1993). These photons can trigger an avalanche in an adjacent µcell.



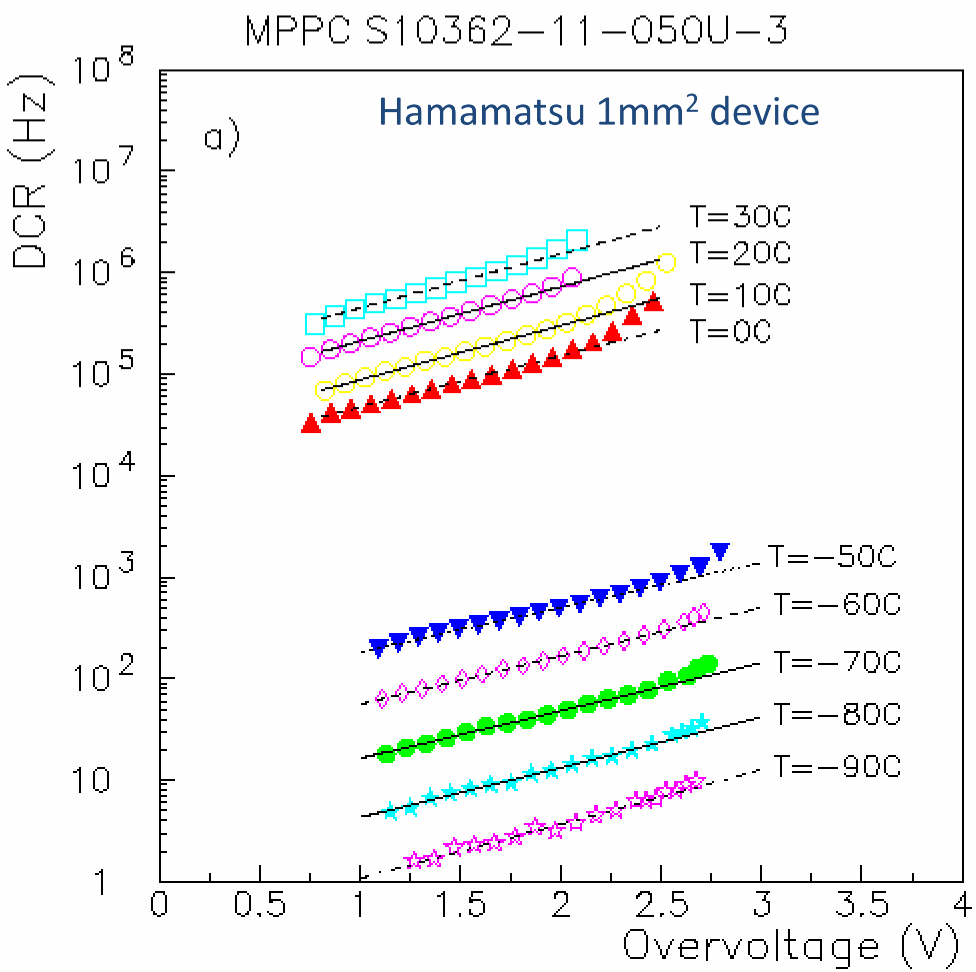
→ Limit gain, increase threshold

→ add trenches btw µcells





... and/or cool



Reduce noise by a factor ~2 every 8 K

Overvoltage must take into account the T-dependence of V_{BD}

N. Dinu et al., NSS *Conf Record (NSS/MIC), 2010 IEEE* , vol., no., pp.215-219, SiPM

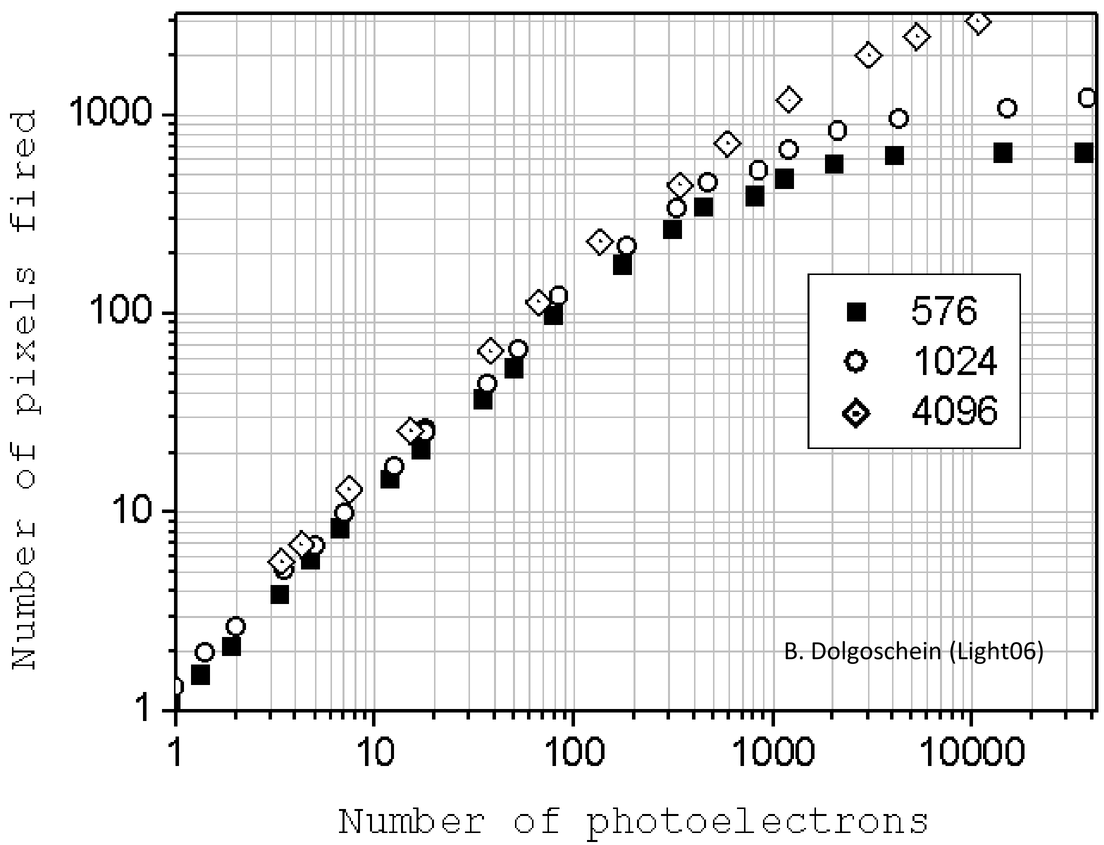
AX-PET

Limited linearity

is simply a consequence of the limited of available pixels. Some pixels get ≥2 hits.

$$N_{fired} = N_{available}\left(1 - \exp\left(-\frac{N_{\gamma,\,true}}{N_{available}}\right)\right)$$

$N_{true}/N_{available}$ ~ 0.5
→ N_{fired}/N_{true} ~ 0.8



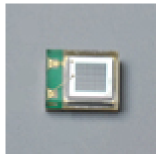
SiPM designs (just examples)

Hamamatsu HPK (http://jp.hamamatsu.com/)

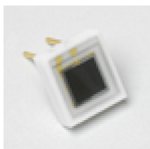
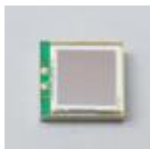
25x25μm², 50x50μm², 100x100μm² pixel size

1x1mm²



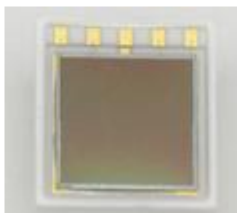


3x3mm²

Arrays

1x4mm²
1x4 channels

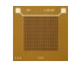


6x6 mm²
2x2 channels

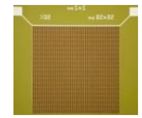
FBK-IRST

50x50μm² pixel size

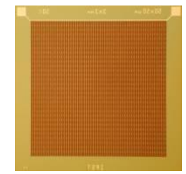
1x1mm²



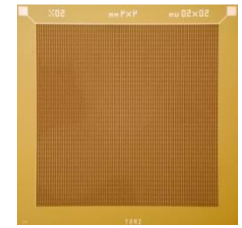
2x2mm²



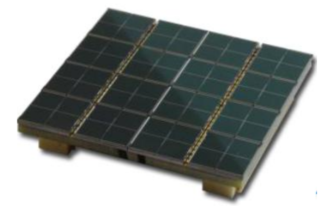
3x3mm²



4x4mm²



4x4mm²
2x2 channels



3x3 cm²
8x8 channels

SensL (http://sensl.com/)

20x20μm², 35x35μm², 50x50μm², 100x100μm² pixel size

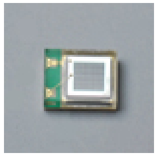
3.16x3.16mm²
4x4 channels

3.16x3.16mm²
4x4 channels

6 x 6 cm²
16x16 channels

PM vs SiPM

	PMT	SiPM
QE (VIS)	0.2-0.4	0.2-0.7
Gain	10^6 @ O(kV)	10^6 @ O(50V)
Timing	T_r ~ O(1ns) TTS ~ O(100 ps)	T_r ~ O(1ns) TTS O(100 ps)
Dynamic range	O(10^6)	O(10^3)
ENF	1.1-1.5	~1
Dark noise rate	O(kHz/cm^2)	O(MHz/mm^2)
Single photon sensitivity/ counting	😐/☹	☺/☺
Magnetic field immunity	☹	☺
Robustness & compactness	☹☹	☺☺



AX-PET

A new geometrical concept for a PET scanner with SiPM readout.

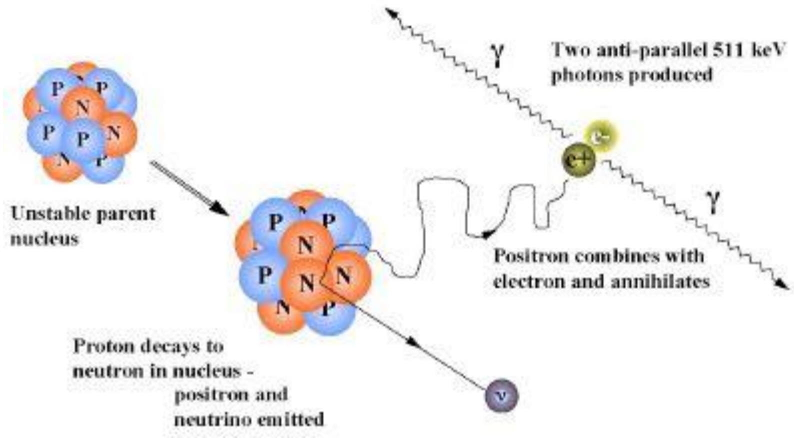
P. Beltrame et al., The AX-PET demonstrator—Design, construction and characterization, NIM A 654 (2011) 546-559
P. Beltrame et al., The AX-PET Concept: New Developments And Tomographic Imaging, 2011 IEEE NSS Conf. Rec. MIC22-5

PET basics

- **Positron Emission :** $p \rightarrow n + e^{+} \nu_e$

β⁺ decay of various radionuclides

Radionuclide	Half life	E_{max}_e^{+}(MeV)
^{11}C	20.4 min	0,96
^{15}O	122 sec	1,73
^{18}F	109,8 min	0,63
^{22}Na	2.6 years	0,55



- **Positron Annihilation :** $e^{+}e^{-} \rightarrow \gamma\gamma$ (E_{γ} = 511 keV). 2 photons emitted almost "back-to-back"

The above two processes define the **fundamental limits** to the spatial resolution of PET

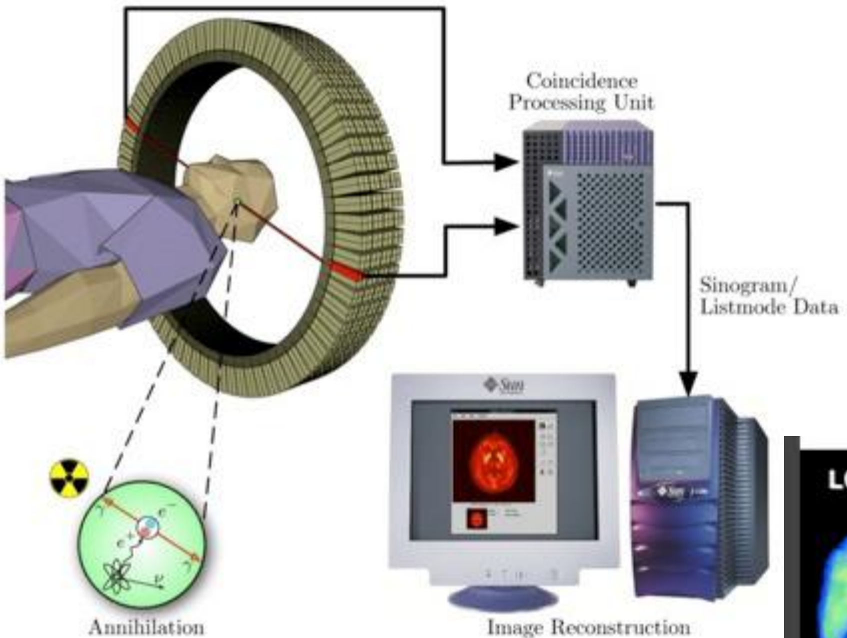
- **Finite positron range (ρ):** annihilation position ≠ emission point, ρ depends on the energy of the positron (i.e. on the radioisotope). **ρ ~ mm**
- **Non-collinearity of the 2 photons:** residual momentum of the $e^{+}e^{-}$ at the annihilation => the 2 photons are emitted with a small deviation from 180° (Δθ ~ 0.5°) => blurring of the spatial resolution **R_{FWHM} ~ 0.0022 x D [mm]**

PET detector principle :

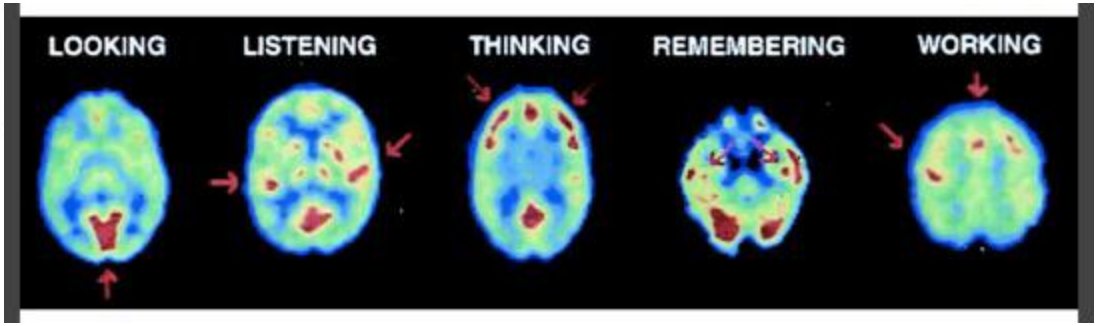
- coincidence of two 511 keV photons define a line of record.
- Take projections under all angle
- (2/3D) Tomographic reconstruct of data

In practice…

1. Inject the radiotracer into the body (0.1-0.2 mCi per kg body weight)
2. Wait for up-taking period (~h)
3. Acquire data (~0.5 h for full body)
4. Feed the data into the reconstruction algorithms
5. (2/3D) image of the activity concentration



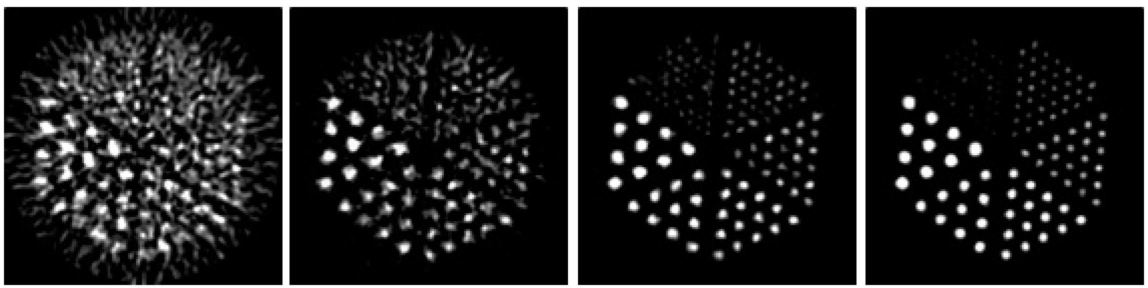
PET is an **"in-vivo" functional imaging** technique
It produces a (quantitative) image of the radio-tracer concentration
Use of F-18 labeled glucose → image of the bodies metabolism



Quality of a PET image depends on

- Detector resolution → crystal size and readout type)
- Counting statistics → efficiency, detector thickness, scintillator material
- Scatter fraction (Compton scattering in body and detector) → energy resolution
- Accidental coincidences and pile-up → coincidence window
- …

Statistics matters ….



A.Del Guerra, CERN
Academic Training 2009

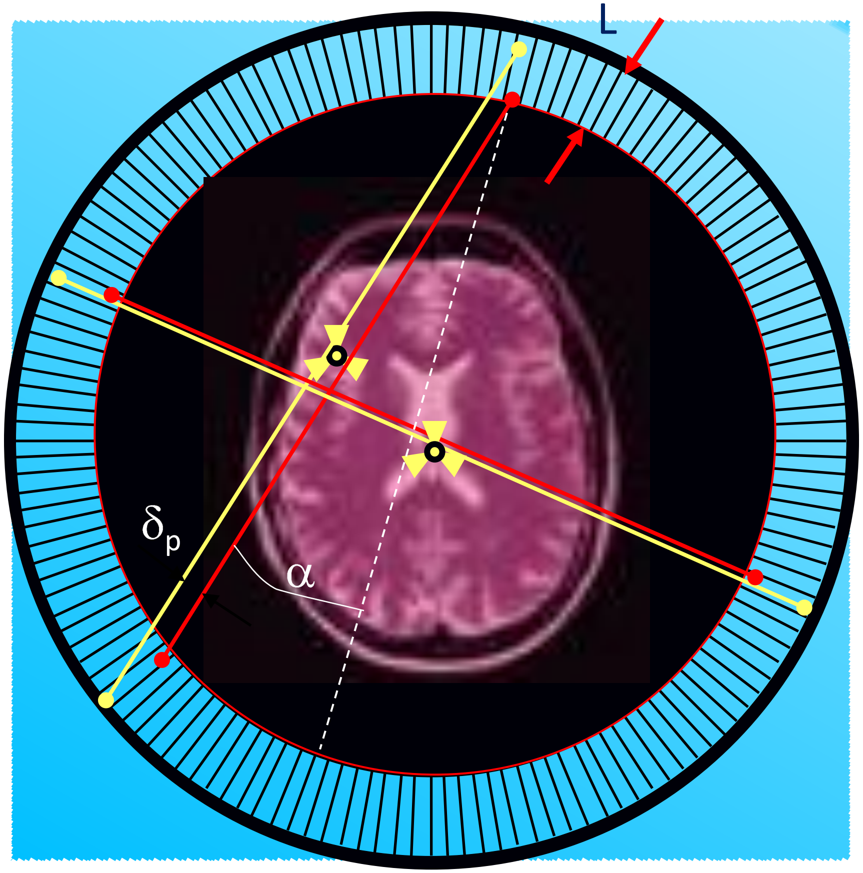
50000 counts | 100000 counts | 500000 counts | 2M counts

The challenges in PET are today (and since a long time)

- Improve resolution (approach physical limits)
- Improve efficiency
- Reduce background
- Combine it with other imaging modalities, in particular MRI

The parallax dilemma

The Axial PET concept was conceived to simultaneously optimize resolution and efficiency.



"Standard" radial PET geometry

- Crystals need to have a minimum thickness L

  Efficiency for pair detection $\varepsilon_2 = \left(1 - e^{-L/\lambda_a}\right)^2$

  λ_a = photon attenuation length of crystal

  $L = \lambda_a \rightarrow \varepsilon_2 \sim 40\%$, $L = 2\,\lambda_a \rightarrow \varepsilon_2 \sim 75\%$, …

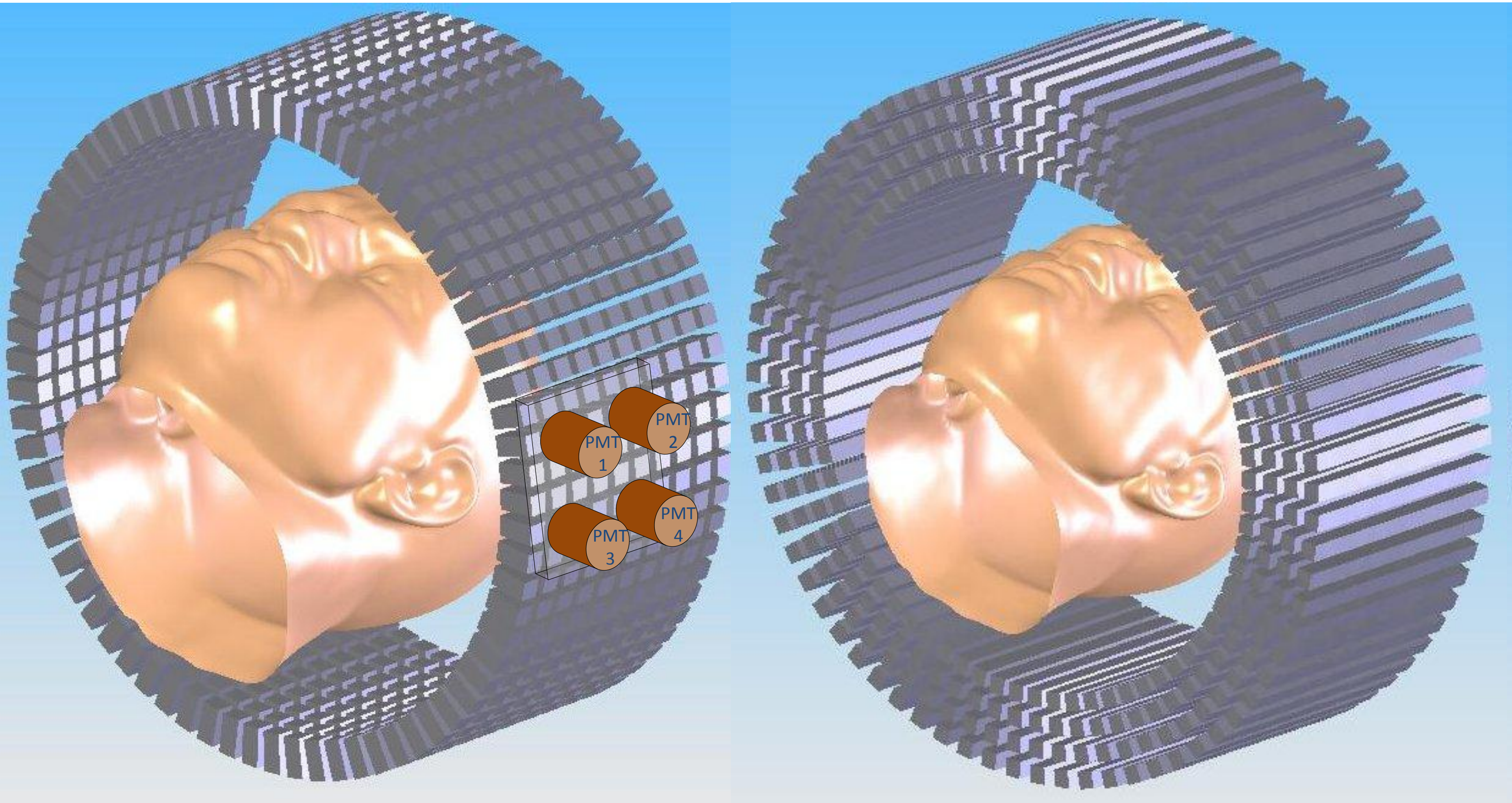
  λ_a = 1-2 cm (depending on material, see below)

- A standard PET does not measure the depth of interaction (DOI) in the crystal.
- This introduces a parallax error $\delta_p = L \cdot \sin\alpha$
- The resolution in the off-center region degrades significantly
- Solution: reduce L ($\rightarrow$bad ε_2) OR measure DOI OR invent a different geometry

The AX-PET concept

SiPM

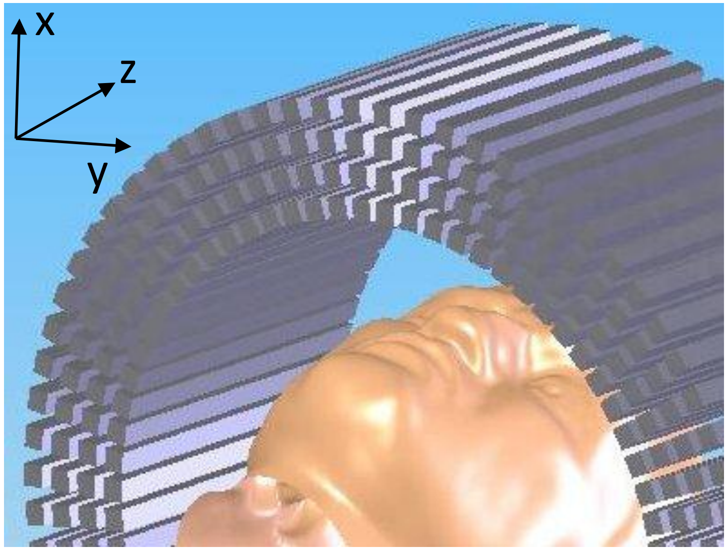
AX-PET

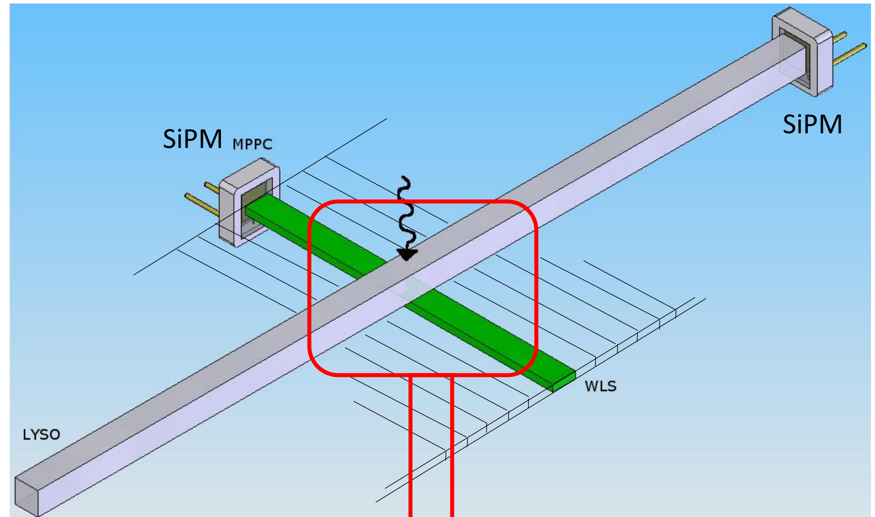


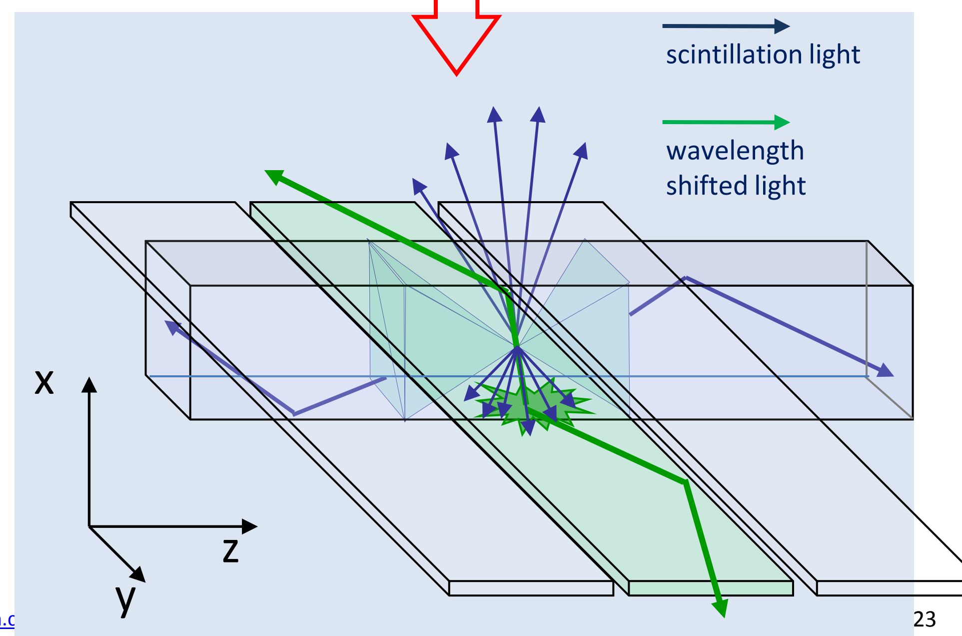
From short radially oriented, block readout crystals …

… to long, axially oriented, individually readout crystals

Our implementation of the AX-PET concept







- Individual readout of crystals. Crystal address gives transverse coordinates (x,y)
- Small crystal cross section → parallax error practically negligible !
- Additional layers allow to increase sensitivity
- → Spatial resolution and sensitivity are decoupled and can be optimized independently

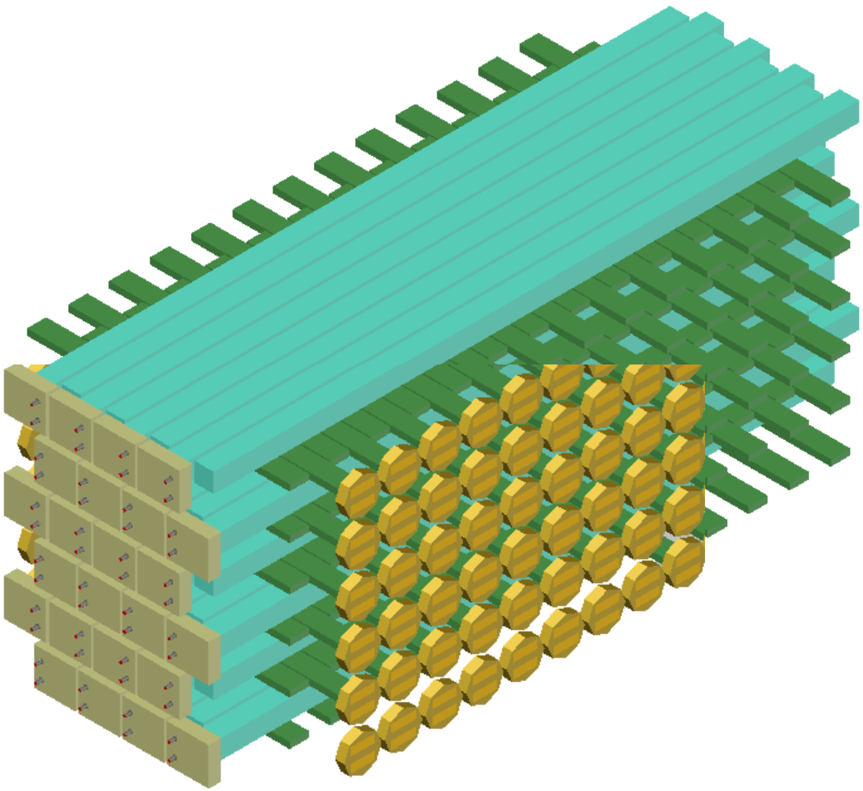
❖ **How to read crystals ?**

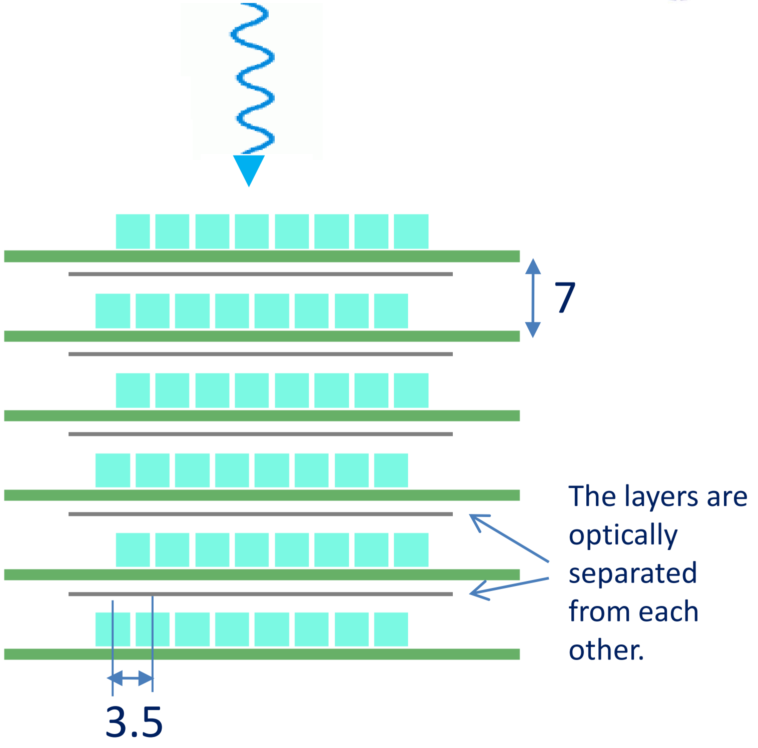
❖ **How to measure axial coordinate ?**

Since 2008, the AX-PET collaboration (Bari, Cagliari, CERN, Michigan, Ohio, Oslo, Tampere, Valencia, Zurich, < 10 FTE) has built and tested a fully operational PET demonstrator scanner.

It consists only of two camera modules à

- 48 crystals (6 layers x 8 crystals)
- 156 WLS strips (6 layers x 26 strips)





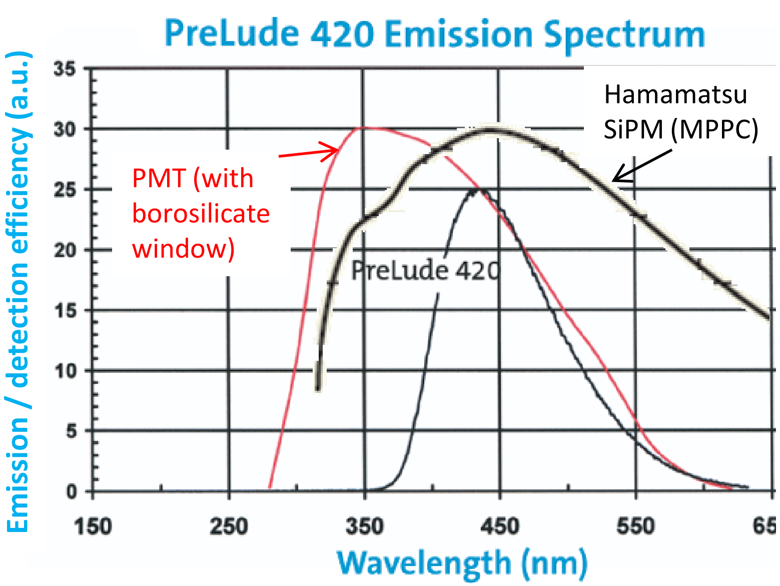
Crystals are staggered by 2 mm. Crystals and WLS strips are read out on alternate sides to allow maximum packing density.

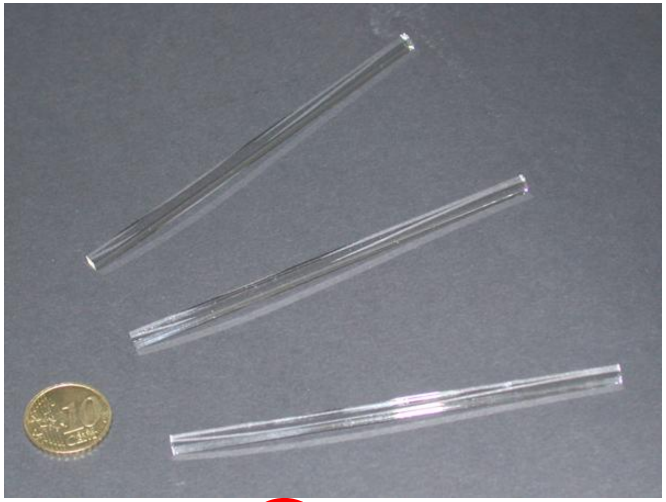
AX-PET components

The **scintillator crystals** are Ce doped LYSO ($Lu_{1.8}Y_{.2}SiO_5$:Ce) single crystals, fabricated by Saint Gobain and commercialized under the trade name PreLude 420.

The main characteristics are:

Density [g/cm3]	7.1
Attenuation length for 511 keV [cm]	1.2
Wavelength of maximum emission [nm]	420
Refractive index at W.L. of max. emission	1.81
Light yield [photons/keV]	32
Average temperature coefficient [%/K]	-0.28
Decay time [ns]	41
Intrinsic energy resolution [%, FWHM]	~8
Natural radioactivity [Bq/cm^3]	~300
Effective optical absorption length [mm]	~ 420





One end is read out, the other end is mirror-coated (evaporated Al-film).

Dimensions: 3 x 3 x 100 mm^3

This determines the transverse resolution (1 module).

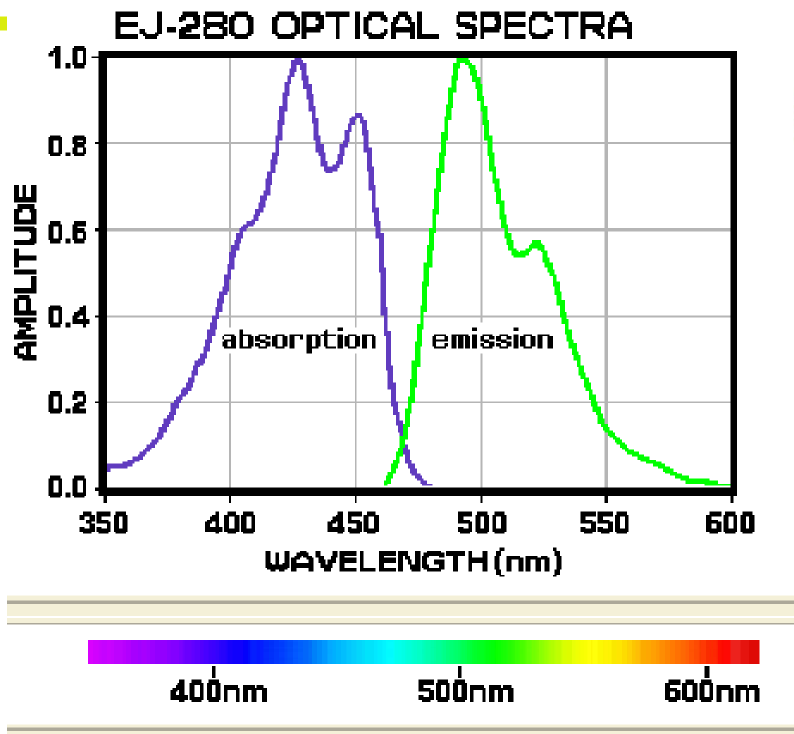
Expect: $\sigma_x = \sigma_y = 3/\mathrm{sqrt}(12) \sim 0.86$ mm (2 mm FWHM)

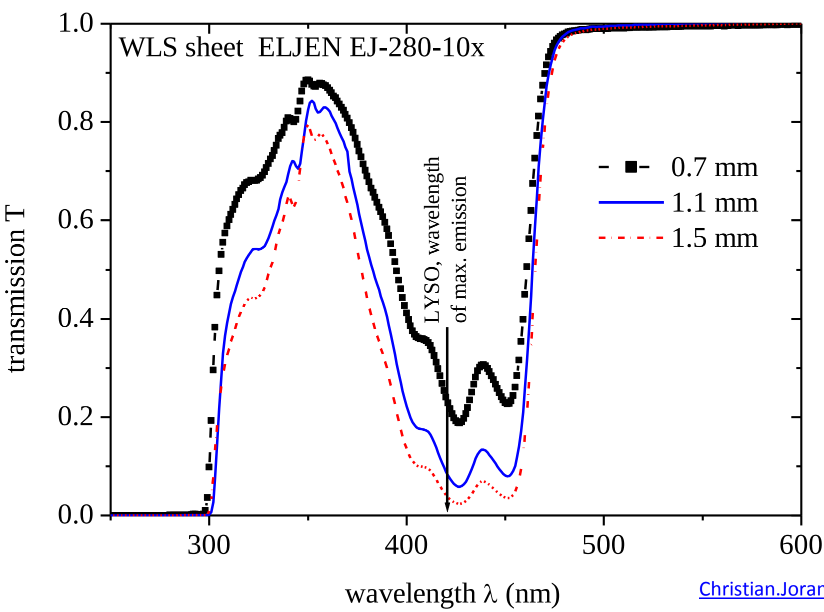
The **WLS strips** are of type EJ-280-10x from Eljen Technologies

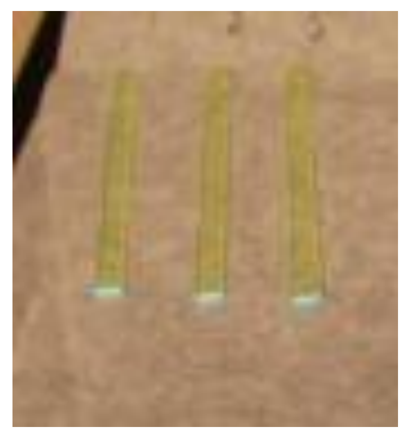
- Shift light from blue to green
- Density: 1.023 g/cm3
- Absorption length for blue light: 0.4mm (10 x standard concentration)
- Index of reflection: 1.58
- Decay time: 8.5ns
- Size: 0.9×3×40mm^3

σ_z = 3/sqrt(12) ~ 0.86 mm (2 mm FWHM).

Use COG algorithm on top.







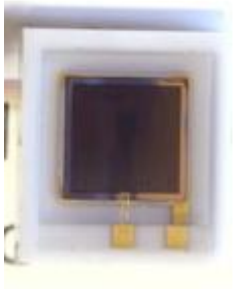
One end is read out, the other end is mirror-coated (evaporated Al-film).

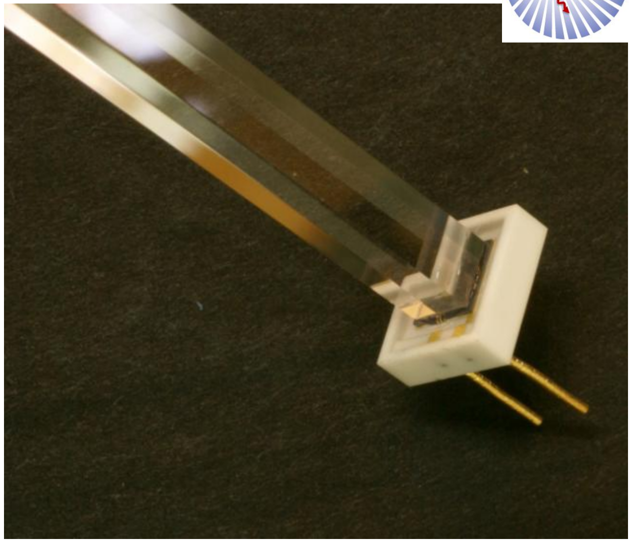
The **photodetectors** are SiPMs of type MPPC from Hamamatsu.

Two different types are used:

MPPC S10362-33-050C → Crystal readout

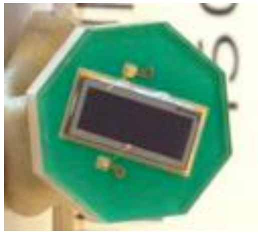
- sensitive area: 3 × 3 mm^2.
- pixel size: 50 × 50 μm^2 (60 × 60 = 3600 pixels)
- typical operational voltage: 70 V.
- Typical gain: 7·10^5.
- Photon detection efficiency: ~40 % (400 nm)





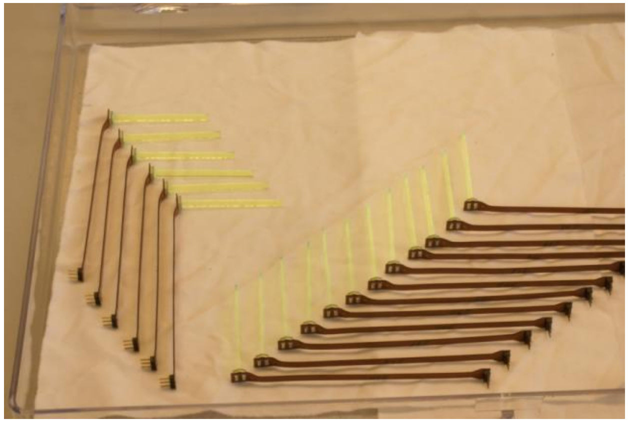
MPPC 3.22×1.19OCTAGON-SMD → WLS readout

- custom specific device
- sensitive area. 3.22 × 1.19 mm^2
- The pixel size is 70 × 70 μm^2,(46 × 17 = 782 pixels).



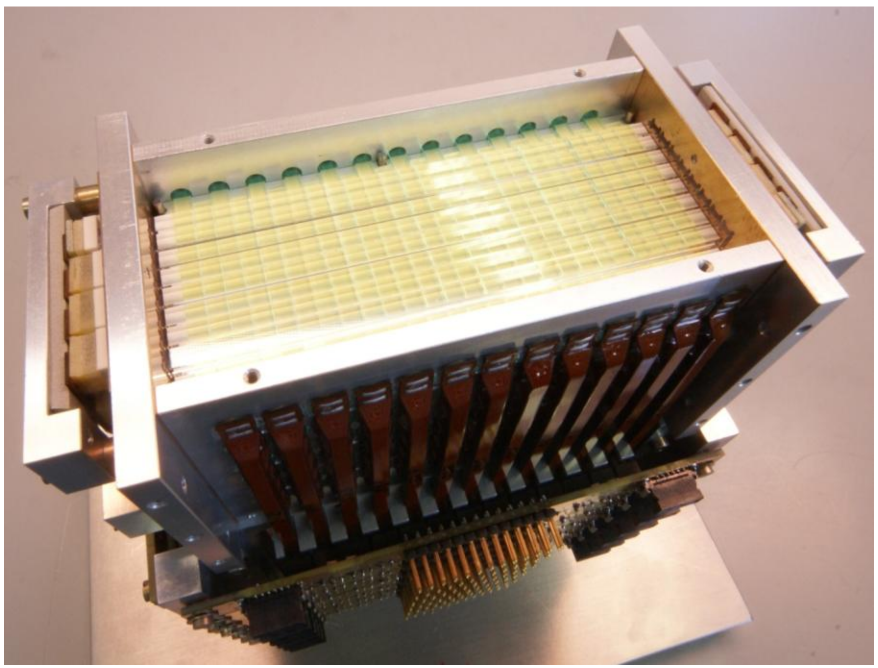
Some MPPC 'features'

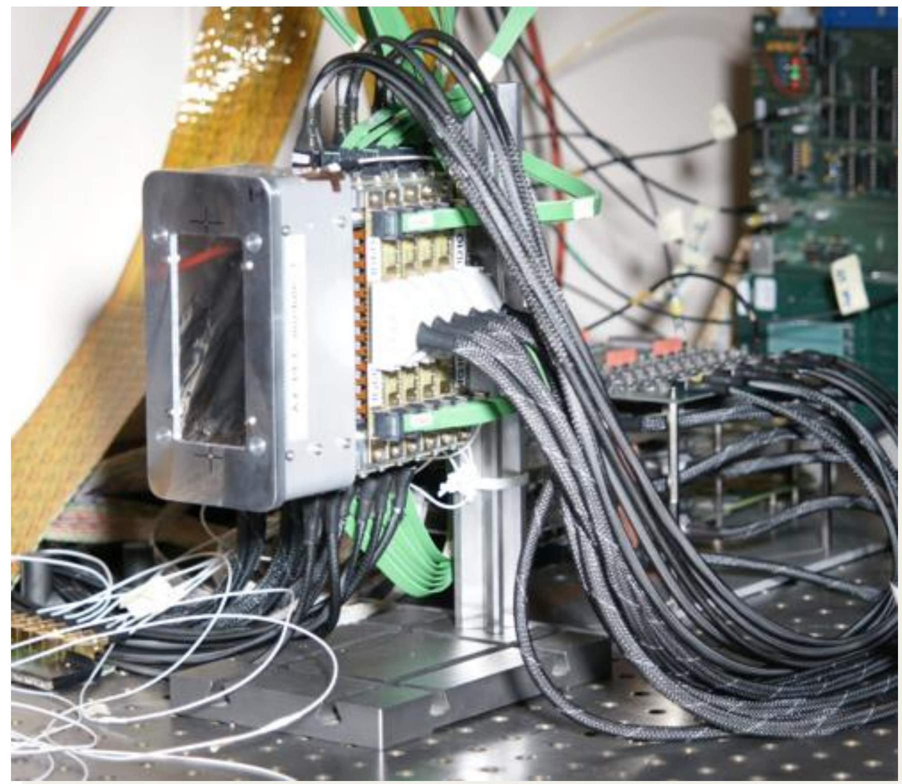
- Gain is temperature dependent (~5%/°K) → T monitoring
- Limited dynamic range → saturation effects → correction
- Thermal noise, O(100 kHz) at 1 photon level. → set threshold



The electrical connections to the MPPCs are made via special Kapton cables soldered to the MPPCs.

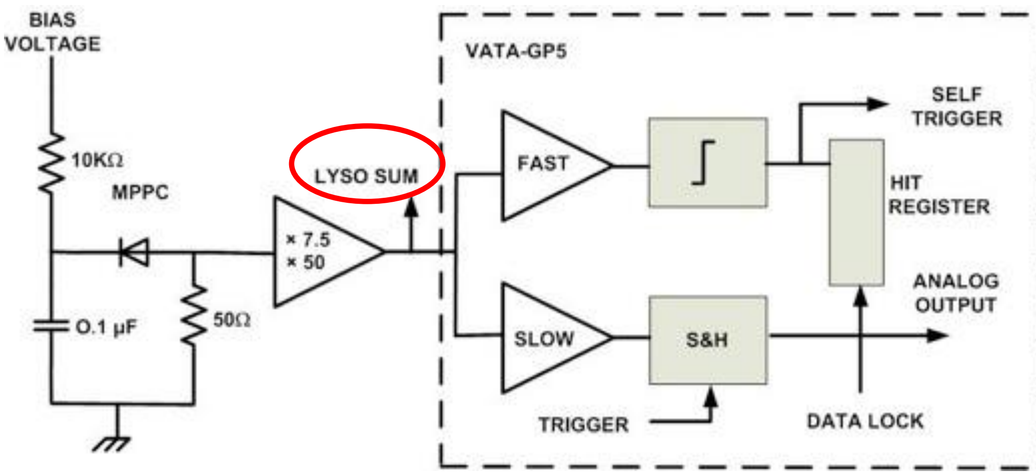
Fully assembled module (48 crystals, 156 WLS stips)
+ lots of cables: 204 x (bias + signal out)

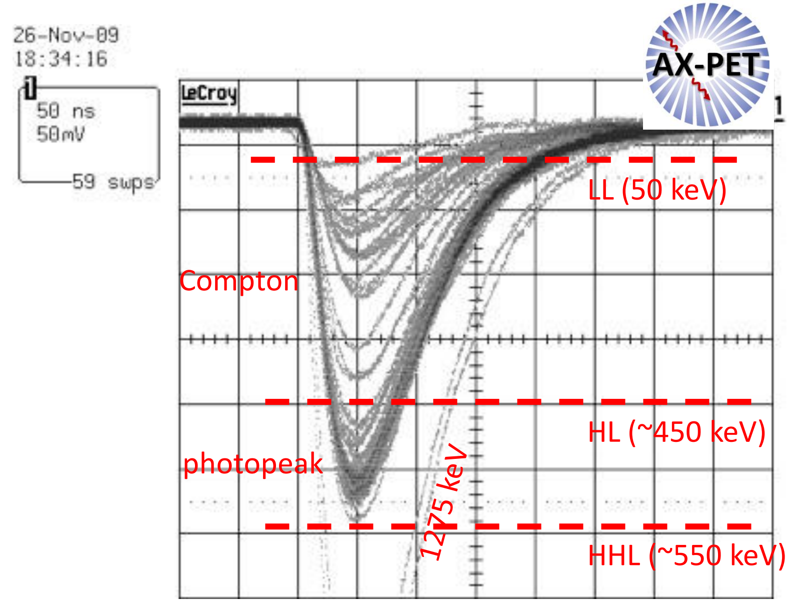




Fully analog, but relatively slow …

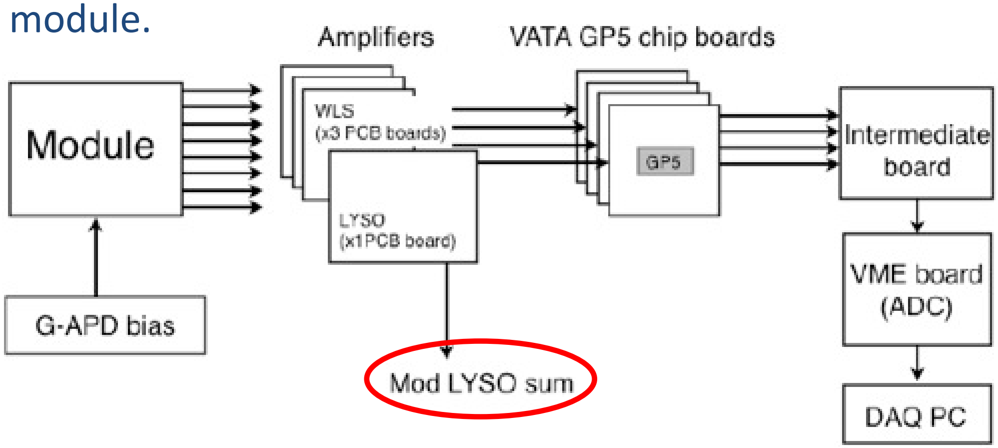
1 channel.



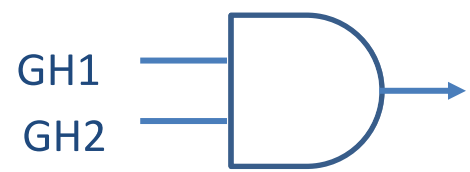


Mod 1: Energy sum of all LYSO crystal of 1 module.
^{22}Na source

1 module.



Good hit in 1 module = LL × HL × $\overline{\text{HHL}}$

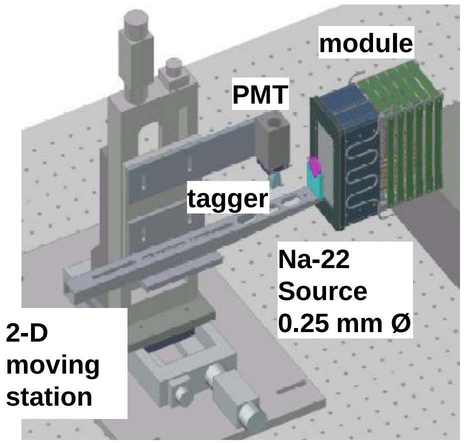


T_{coinc} = 20 ns

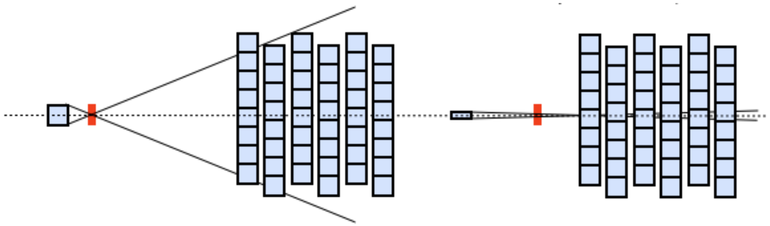
Max Readout rate ~ 20 kHz

Test set-ups

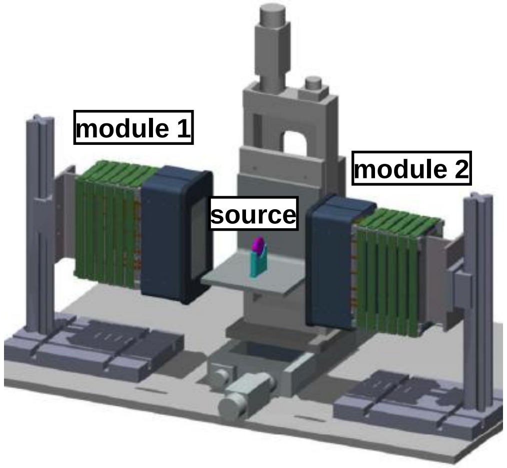
I. Single module characterization



- Module in coincidence with a tagging scintillator
- Use of different tagging crystals

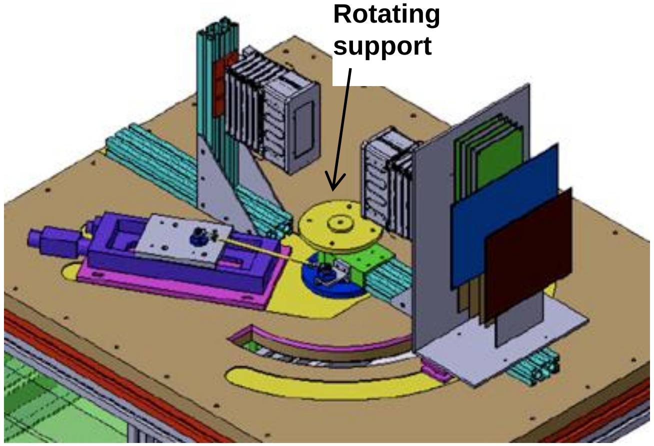


II. Coincidence of 2 modules

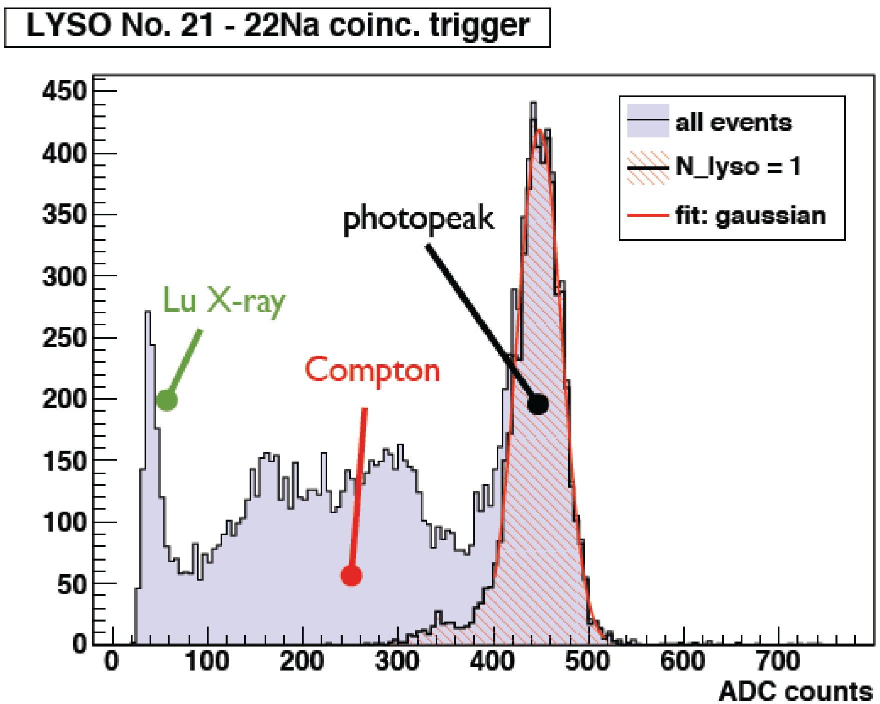


- Distance between modules = 15 cm

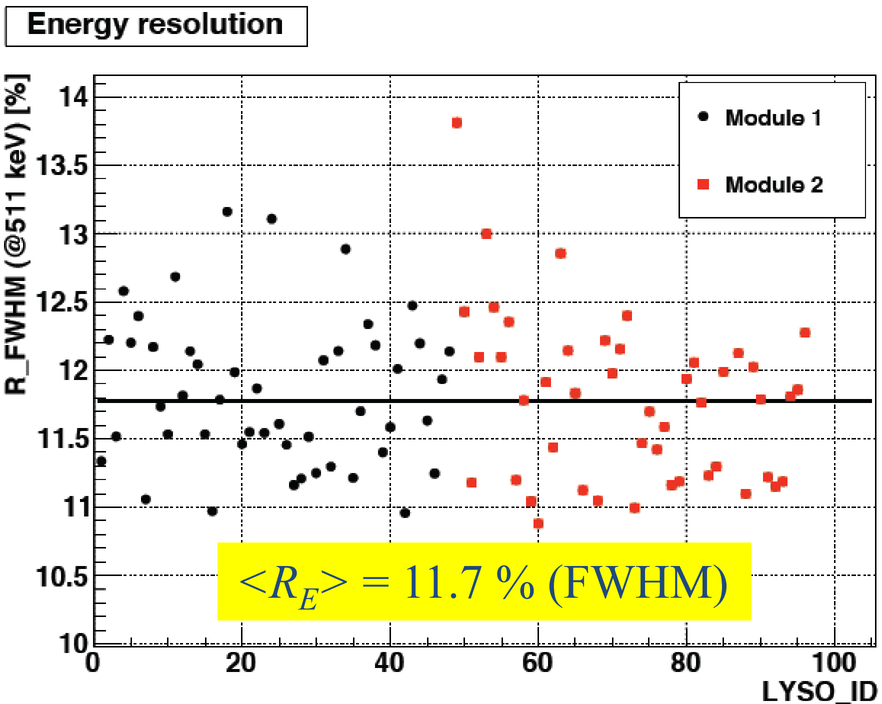
III. Gantry for tomographic measurements

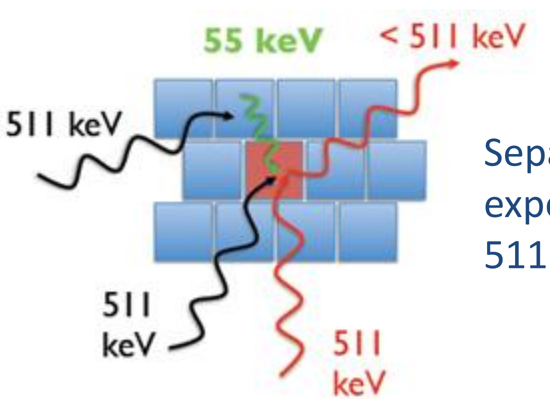


Typical charge spectrum from 1 LYSO crystal



Energy resolution of all 96 LYSO crystals



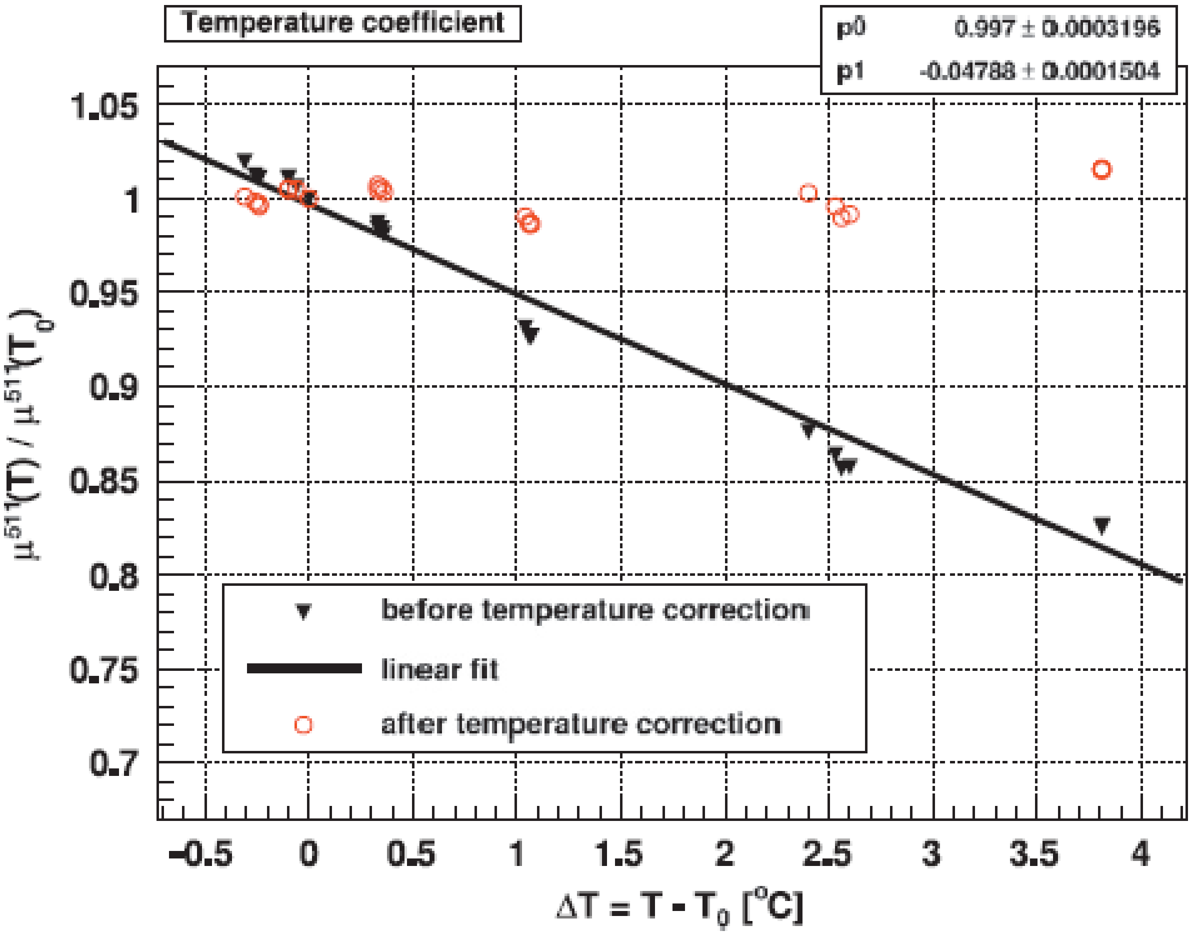


Separate calibration experiment:
511 keV ~ 1100 pe in SiPM

In the ideal case: $R_E = R_E^{\text{intr.}} \oplus \frac{1}{2.35 \cdot \sqrt{N_{pe}}}$

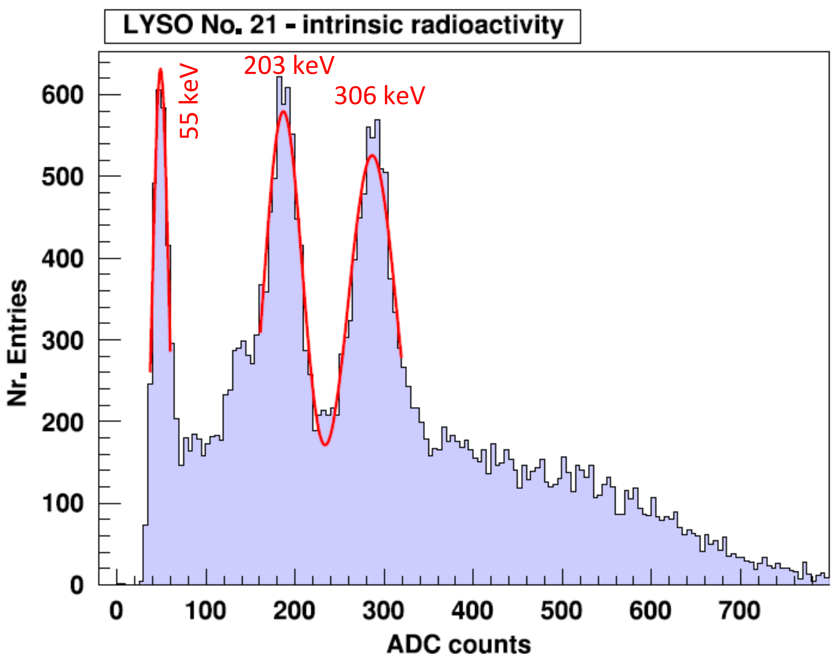
→ Expect R_E ~ 10.7 %

First correct for possible temperature changes
SiPM gain drops by ~5% per degree K.

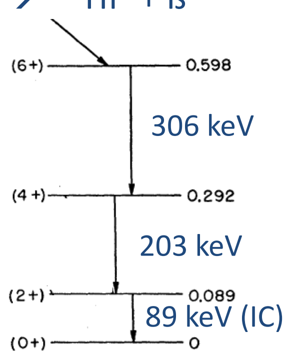


Energy calibration

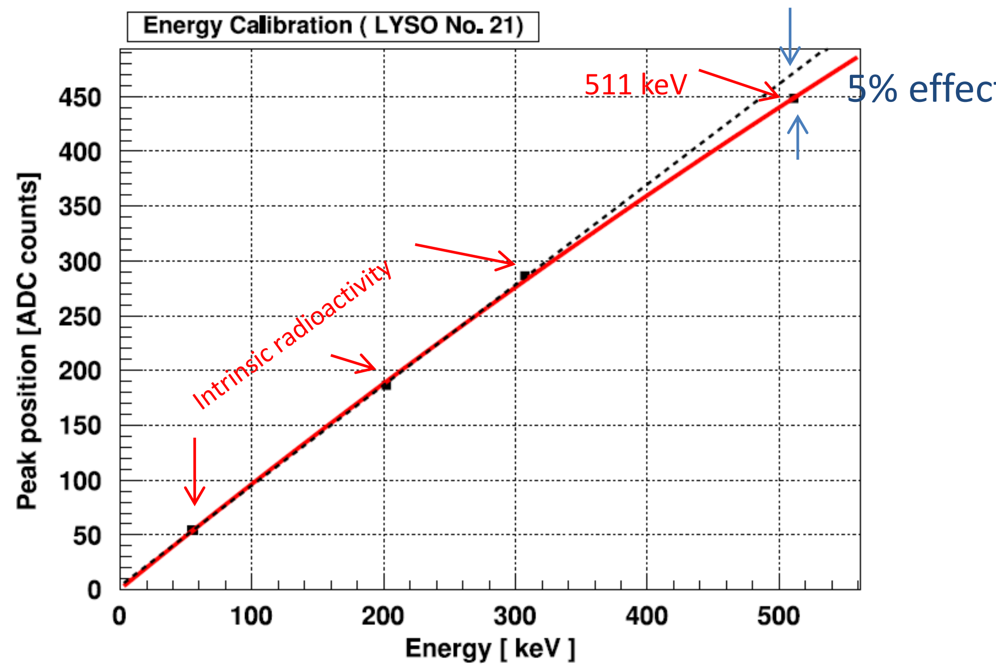
Lutetium in LYSO is radioactive.
~ 250 Bq per crystal.



$^{176}Lu \rightarrow {}^{176}Hf^{*} + \beta^{-}$



Small non-linearity due to limited pixel number of SiPM (3600)

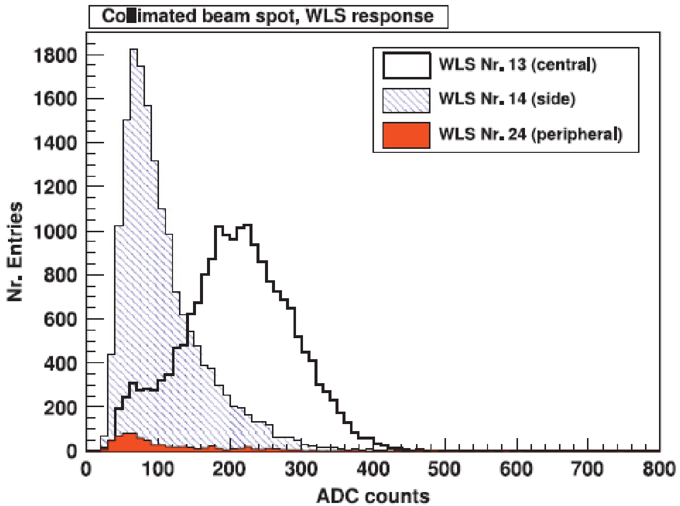


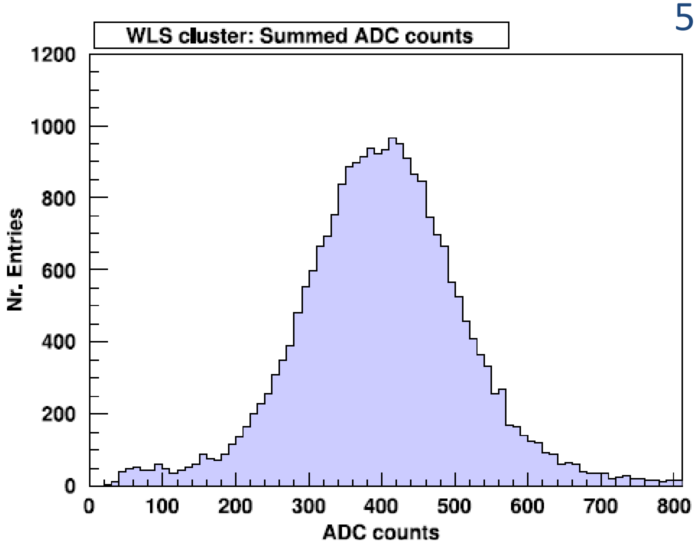
The formula would predict a 15% effect

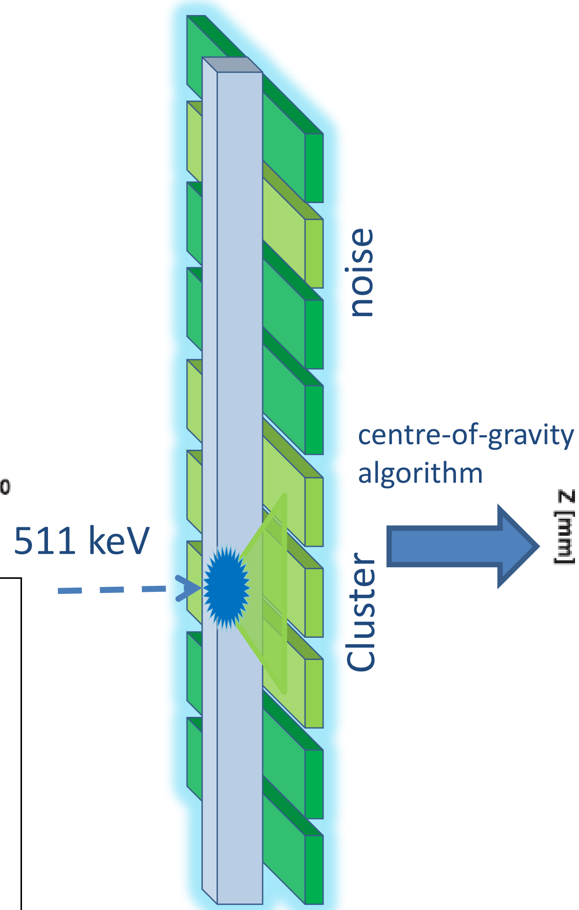
$$N_{fired} = N_{available}\left(1 - \exp\left(-\frac{N_{\gamma,\,true}}{N_{available}}\right)\right)$$

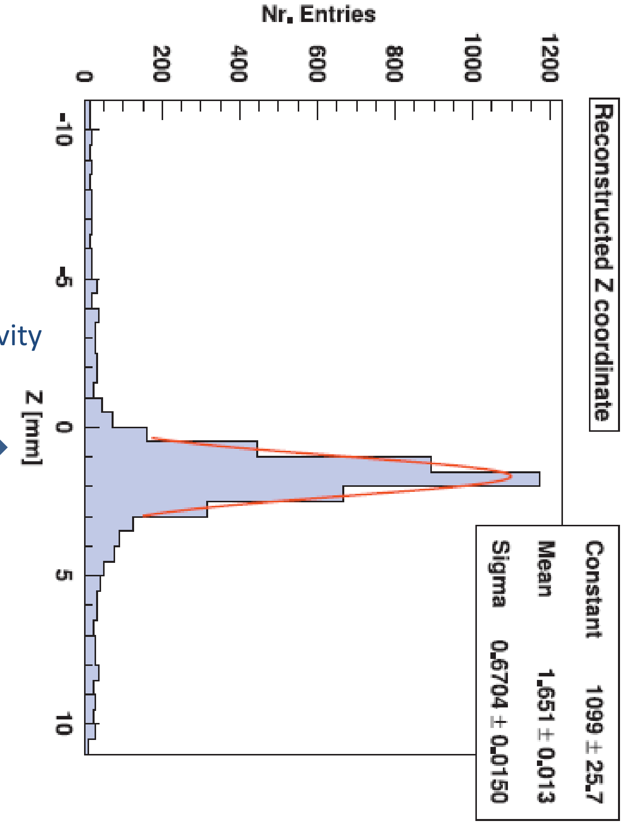
LYSO photons arrive over ~100 ns. Some hit pixels will have recovered when the 2nd hit arrives.

The axial coordinate …









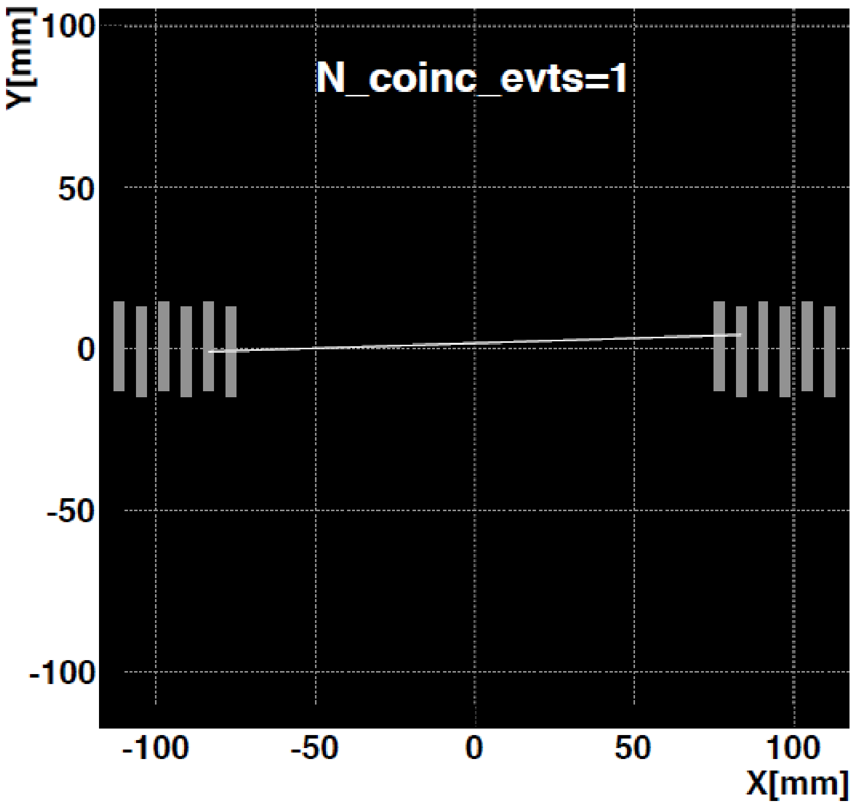
σ_z ~ 0.75 mm
(average over whole module, after correction for e^+ annihilation physics).

Calibration (separate set-up):
511 keV → 100 pe in WLS cluster

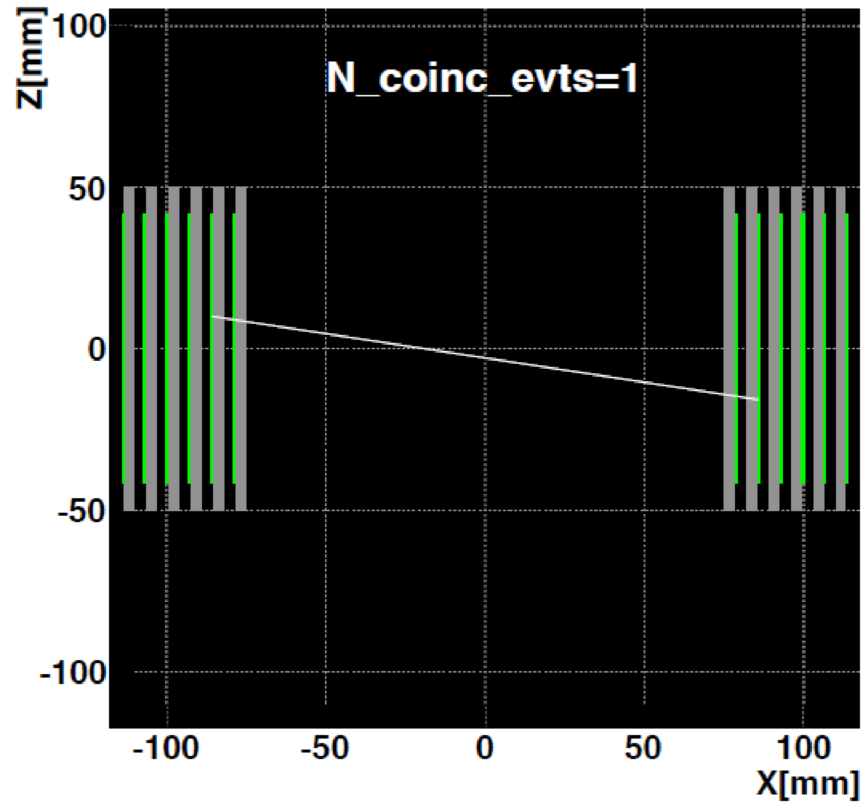
First coincidence measurements

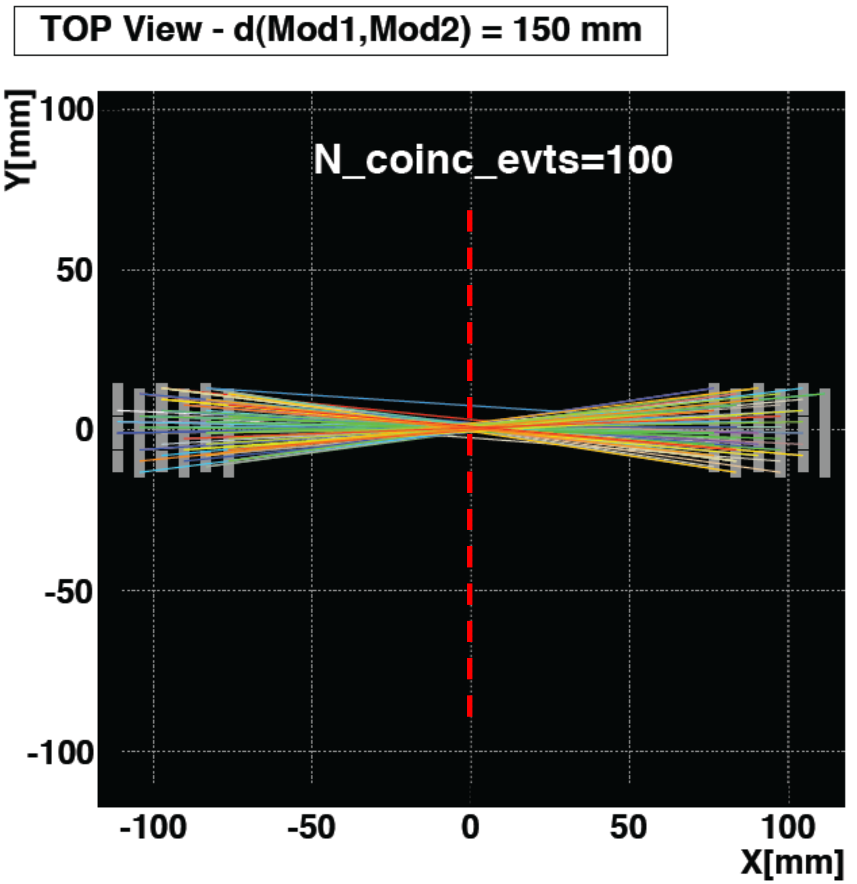
- Point-like ^{22}Na source
- Photoelectric events only (1 hit crystal per module)
- Draw “LOR” (pure geometrical, no tomographic reconstruction)

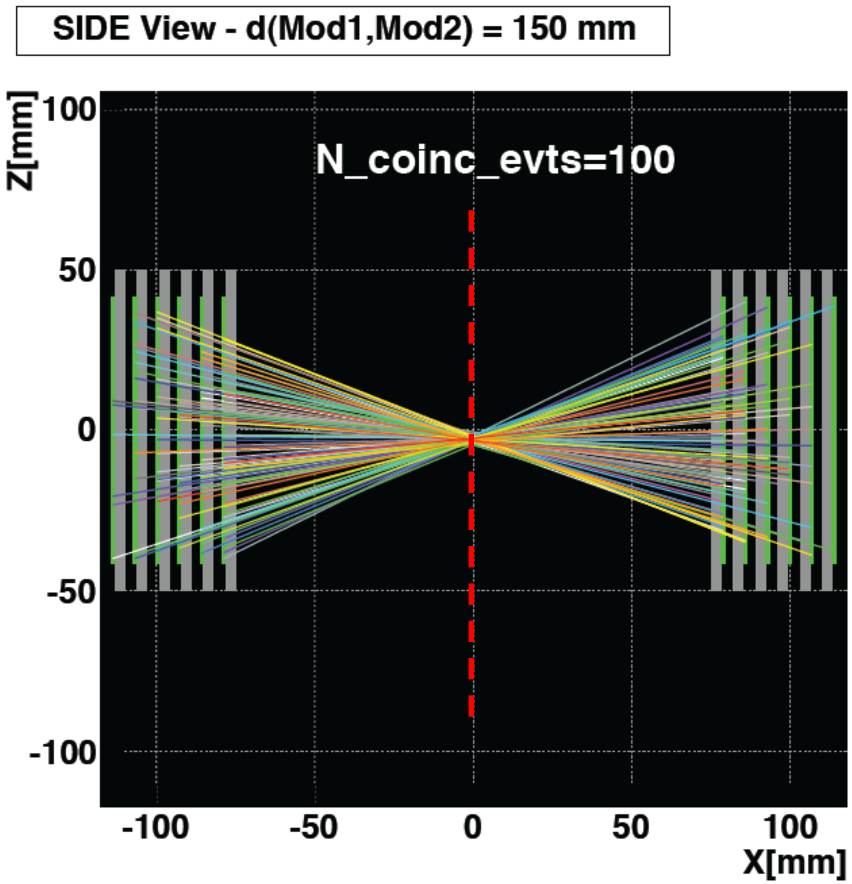
TOP View - d(Mod1,Mod2) = 150 mm



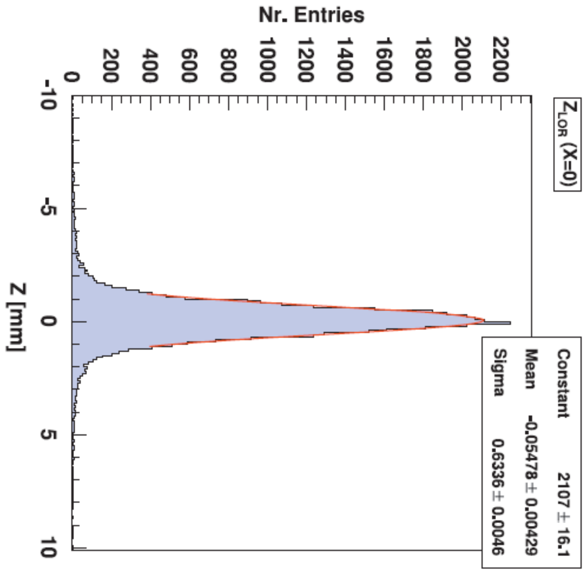
SIDE View - d(Mod1,Mod2) = 150 mm

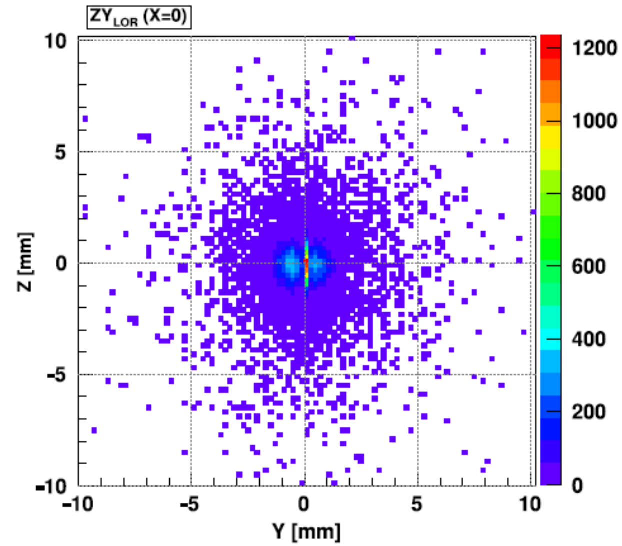


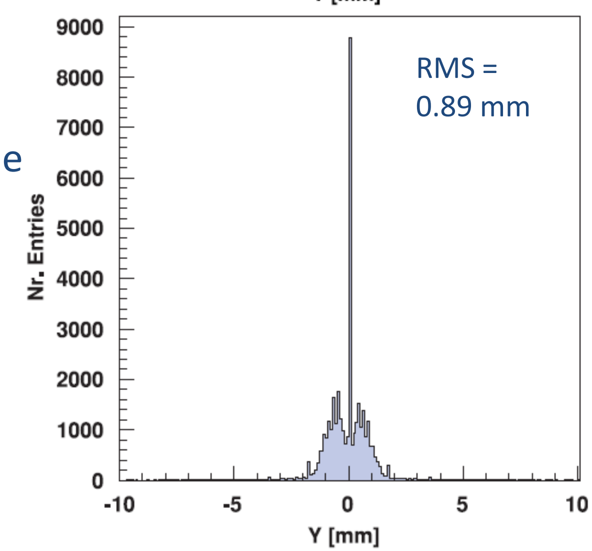

TOP View - d(Mod1,Mod2) = 150 mm
N_coinc_evts=100
Y[mm]
X[mm]
100
50
0
-50
-100
-100
-50
0
50
100


SIDE View - d(Mod1,Mod2) = 150 mm
N_coinc_evts=100
Z[mm]
X[mm]
100
50
0
-50
-100
-100
-50
0
50
100

Look at the YZ distribution at x=0 (“confocal reconstruction”)







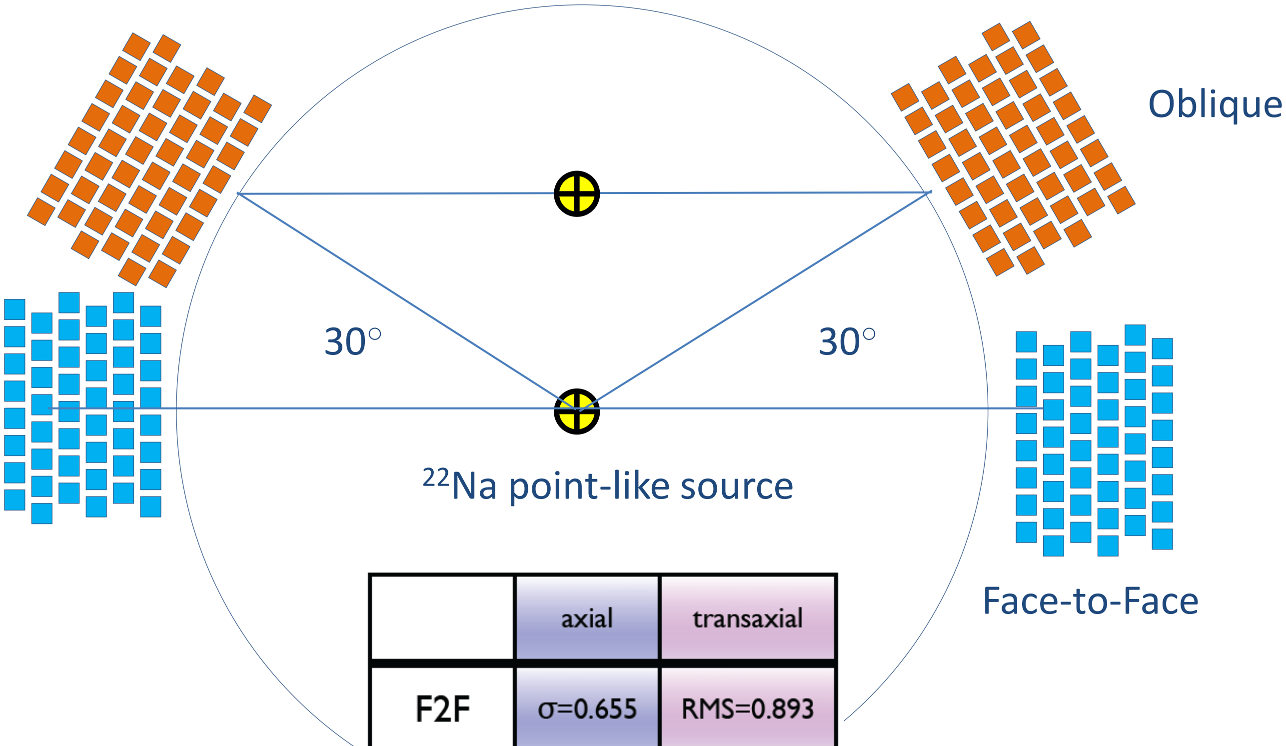
σ_z = 0.63 mm (coincidence) is compatible with the single detector resolution.

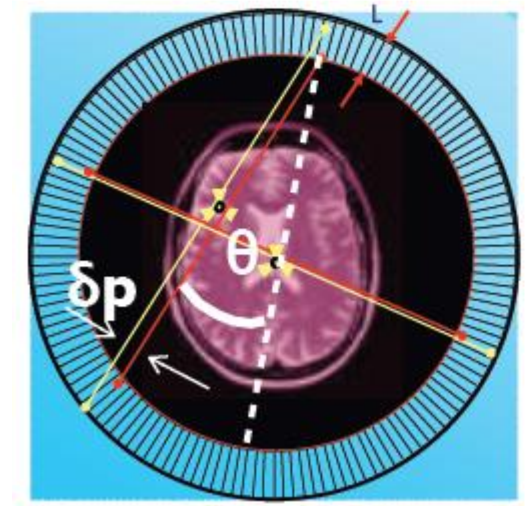
After correction of the e^+ annihilation physics:

σ_z (coinc.) ~ σ_z (detector) / sqrt(2)

y coordinate is quasi-discrete (crystal positions).

Experimental check: Are there parallax errors ?

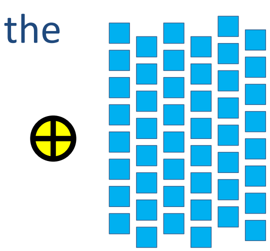




	axial	transaxial
F2F	σ=0.655	RMS=0.893
OBL	σ=0.643	RMS=0.909

As expected from the design, there is no noticeable parallax error, neither in the axial nor transaxial plane.

There is however an unavoidable degradation of the resolution, when the source is very close to one detector. The coincidence resolution looses the 1/sqrt(2) 'bonus'



Basics of tomographic imaging (in AX-PET this is the speciality of the team from Valencia (M. Rafecas et al.)

A PET scanner measures projections of the activity distribution. Reconstruction means to calculate from these projections the actual 3D distribution of the activity. There are two groups of methods:

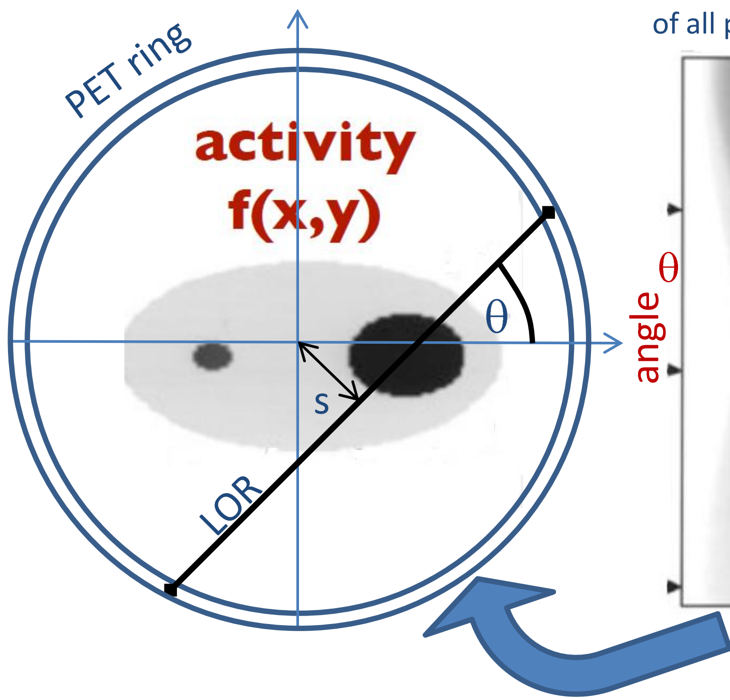
1. Analytical algorithms

   (usually based on sinograms)

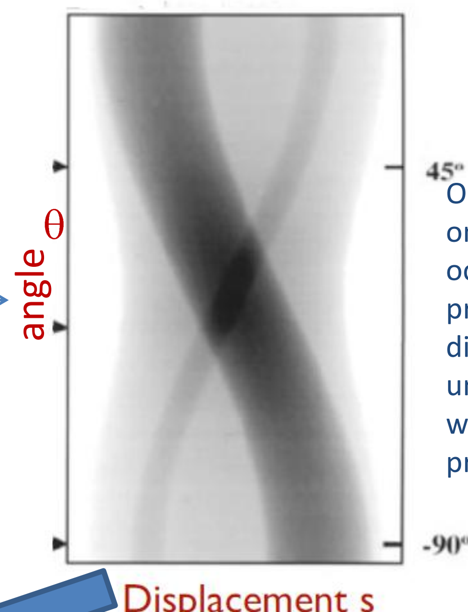
2. Iterative algorithms

   (mainly used in AX-PET, see back-up slides)

(2D case)

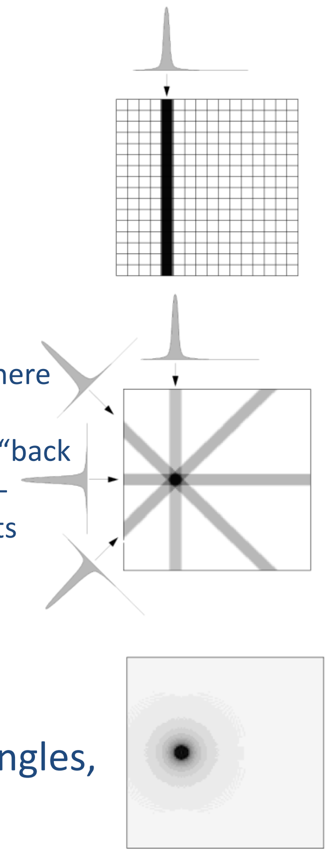


"sinogram"
a 2D histogram
of all projections g(s,θ)



(Filtered) Back Projection

One doesn't know where on the LOR the count occurred. Therefore, "back projection" means re-distributing the counts uniformly to all pixels which fall within the projection path.



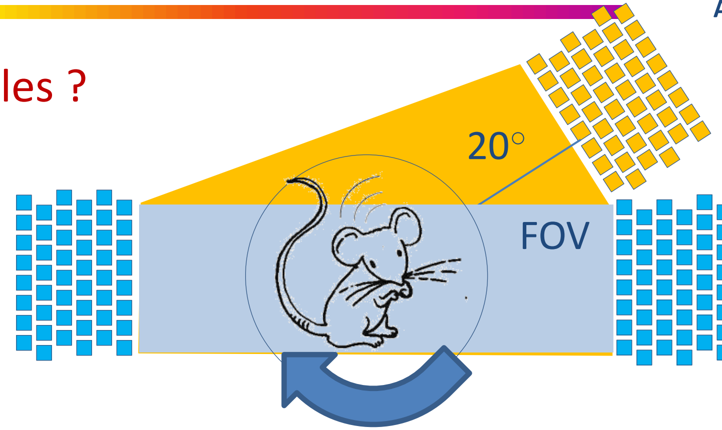
1 angle

4 angles

64 angles

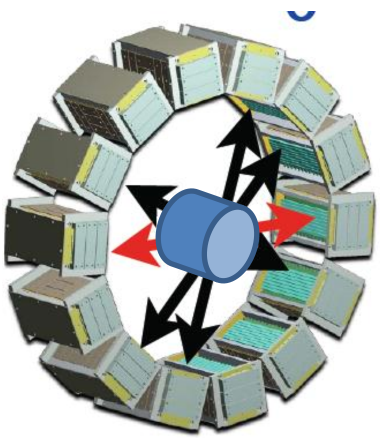
The more angles, the better!

How do we get all these angles with just 2 modules ?

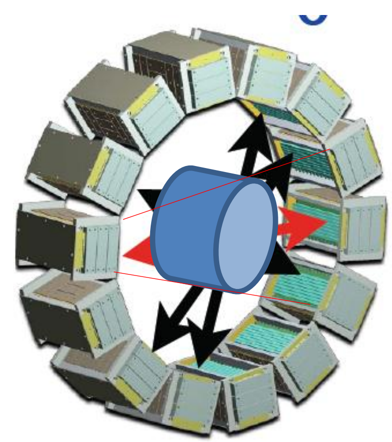


"Step and shoot"
Rotate object in 9 steps of 20 deg.

"Step and shoot"
1.) Rotate object in 9 steps of 20 deg.
2.) Rotate 1 module by 20 deg.
3.) Rotate object in 18 steps of 20 deg.



Price to pay:
- Longer acquisition times (several h).
- Decaying activity
- Changing conditions (pile-up, dead time)



We mimic a 18-module scanner, however with a very limited FOV (width and length of 1 module). We allow only coincidences between opposite modules.

Now we allow coincidences between opposite ±1 modules. → Larger FOV.

The AX-PET set-up has exactly the size of a EURO palette. Very useful for measurement campaigns at other places

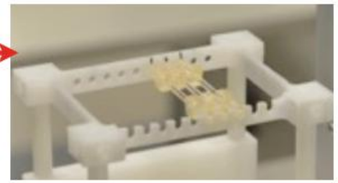
Measurements with phantoms or animals can’t be made at CERN!

ETH Zurich, PET lab, 2010 AAA (Saint GEnis), F) 2010 AAA (Saint GEnis), F) 2011	phantoms
ETH Zurich, PET lab 2012	animals

Phantoms are just containers which can be filled with a radiotracer. We usually work with F18 in aqueous solution. Some ink is added to see air bubbles.

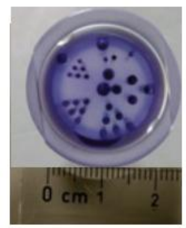
(1) capillaries

L = 3cm ;
Diam = 1.4 mm ;
Pitch = 5 mm



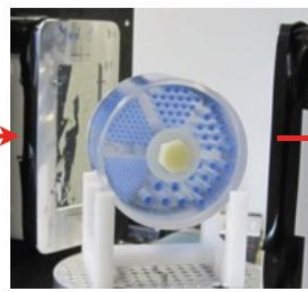
(2) micro Derenzo

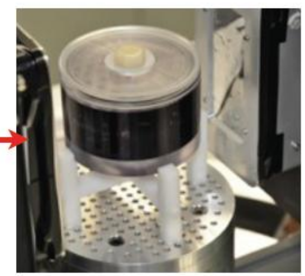
H = 1.5 cm ;
Diam = 2 cm;
Rods_Diam = 0.8÷2 mm



(3) mini DeLuxe

H = 5 cm ;
Diam = 7.5 cm;
Rods_Diam = 1.2÷4 mm





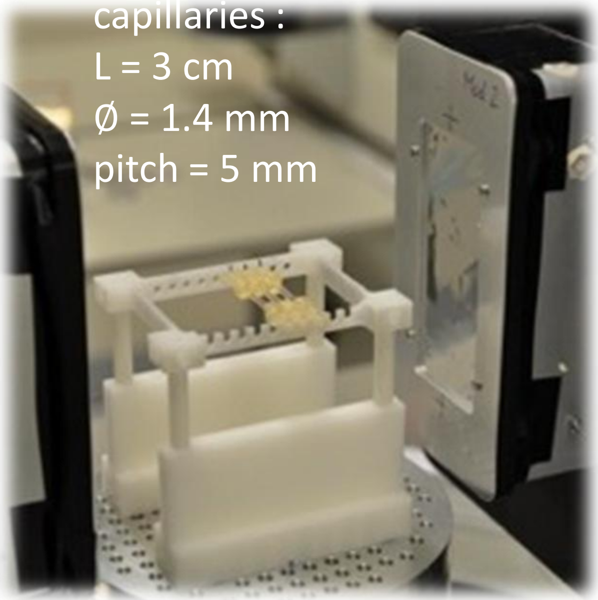
(4) homogeneous cylinder

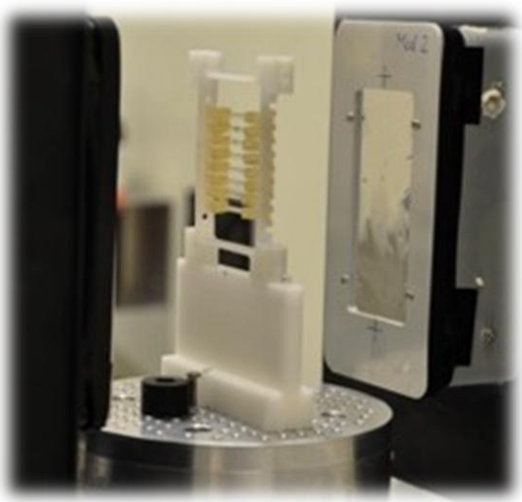
H = 9 cm ;
Diam = 6 cm;

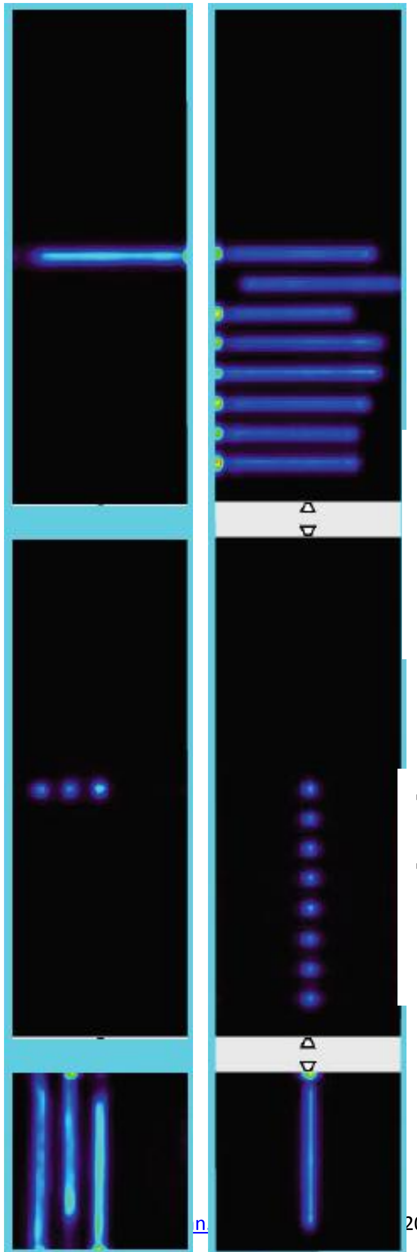
(5) NEMA phantom

H = 6.3 cm ;
Diam = 3.3 cm;

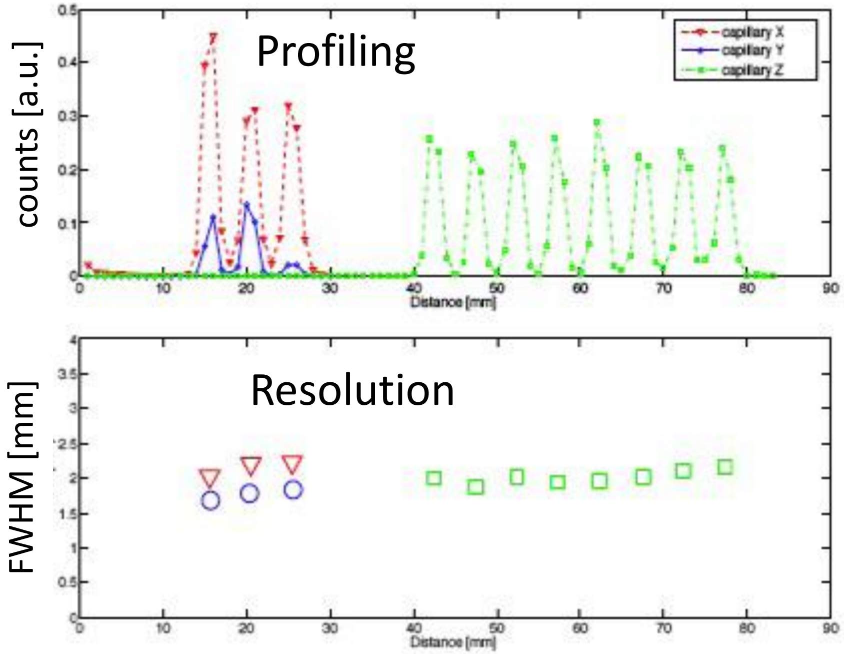
capillaries :
L = 3 cm
Ø = 1.4 mm
pitch = 5 mm



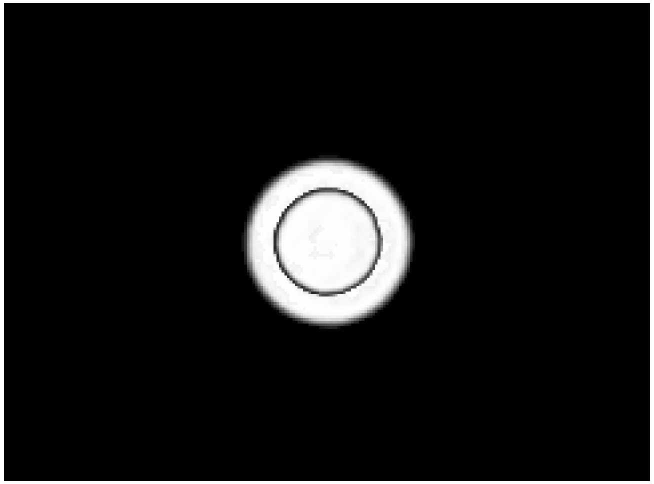




ETH 2010



Three regions in the same phantom to address three different aspects

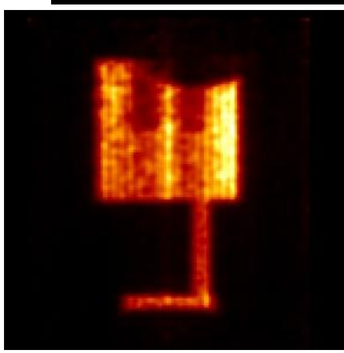


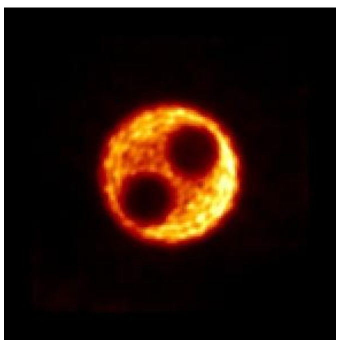
Hot & Cold rods for contrast

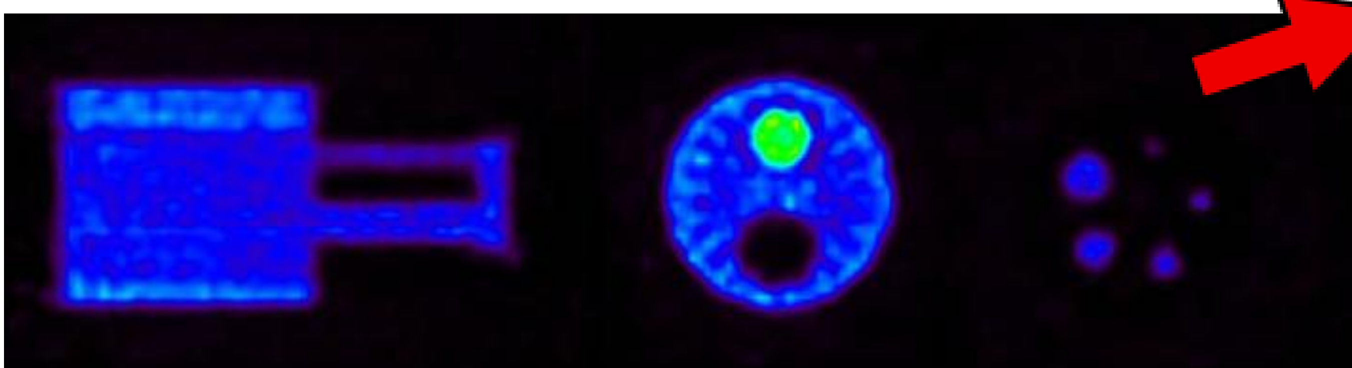
Homogeneous cylinder for assessing the ability to reconstruct homogeneous distributions

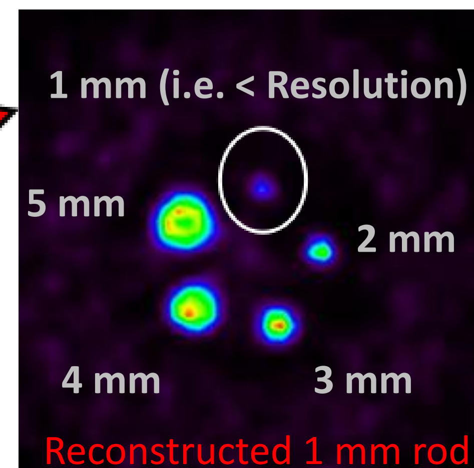
Series of small rods for resolution

NEMA mouse phantom



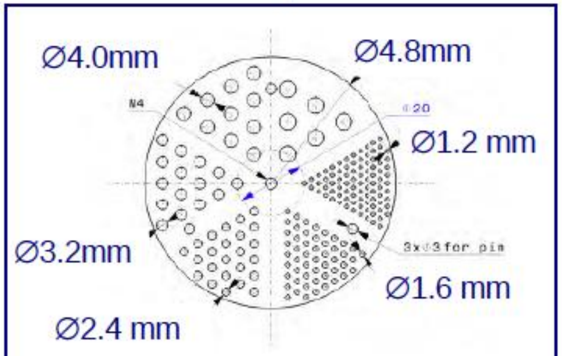






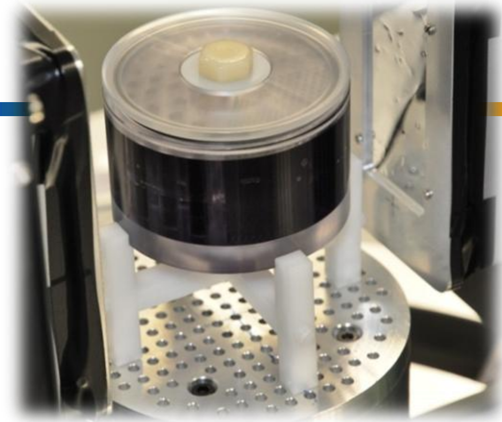
=> FWHM ~ 1.6 mm

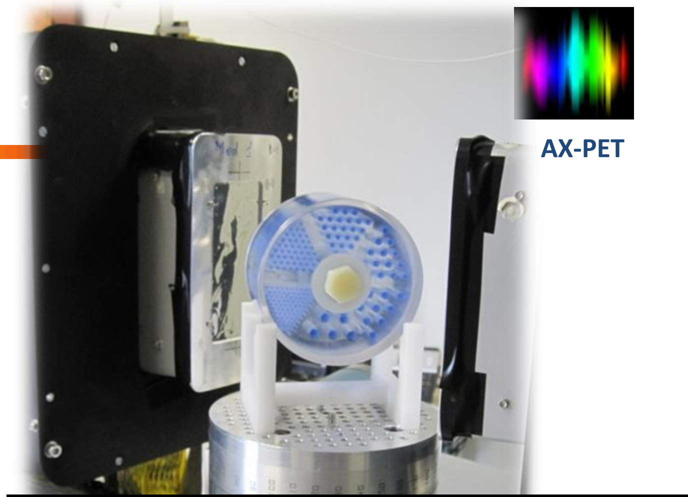
Mini Deluxe phantom



Hollow rod diameters

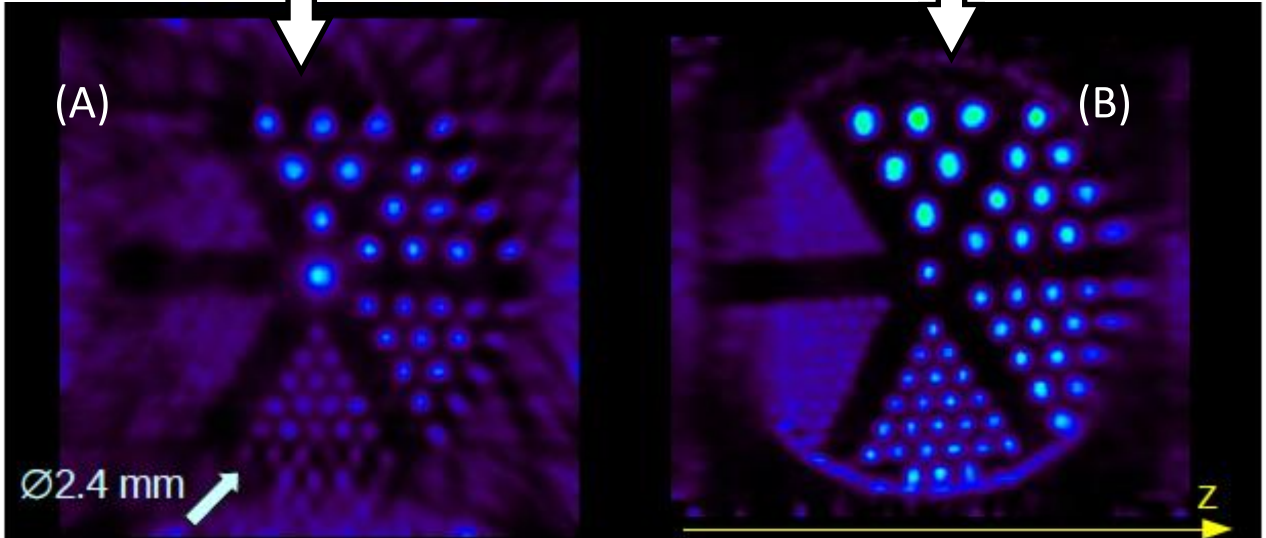
- High density resolution phantom
- Diameter (75 mm) is larger than the extended FOV





Rods oriented parallel to Z axis

Rods oriented perpendicular to Z axis



- Fixed time acquisition: 120 s /step
- 60 iterations + post-reconstruction smoothing
- No corrections
- Artefacts due to data truncation (FOV too small...)

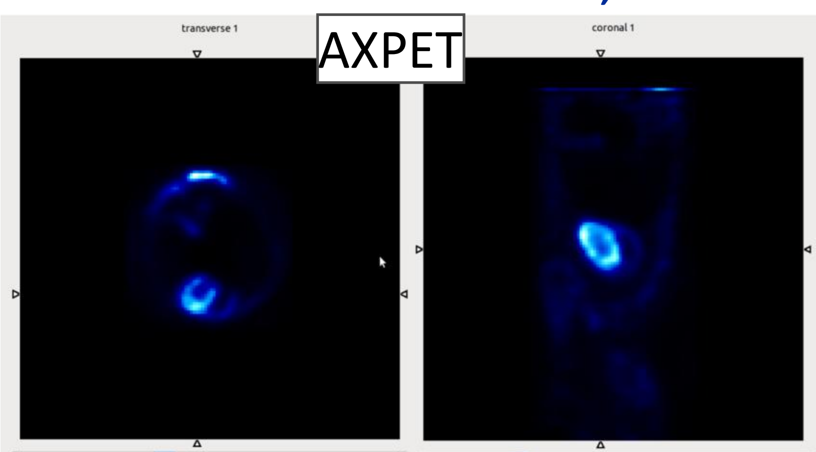
Small Animal Tests

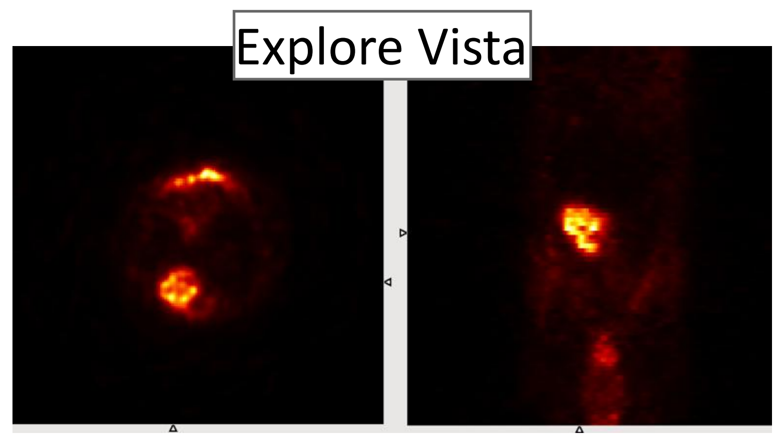
- June 2012, small animals tests campaign at ETH Zurich, Radiopharmaceutical Inst.
- Possibility to use their GE Explore Vista PET/CT scanner as reference. This is a dedicated small animal scanner with 1.45 x 1.45 mm crystals. Block readout. DOI via Phoswich approach.
- one mouse, FDG
- one (young) rat, FDG

=> organs structure

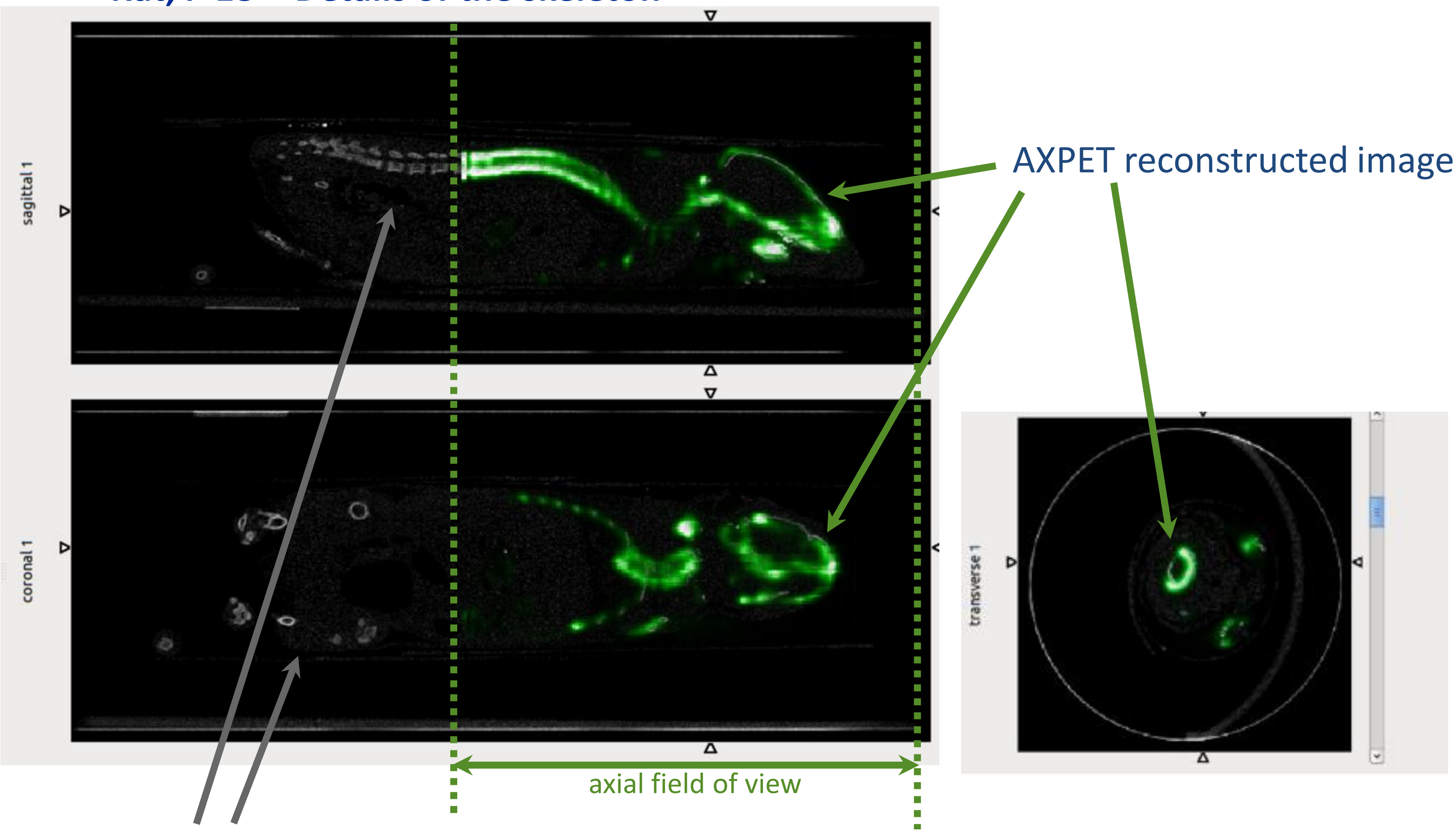
- one (young) rat, 18-F => bones, skeleton structure

Rat, FDG - Details of the heart





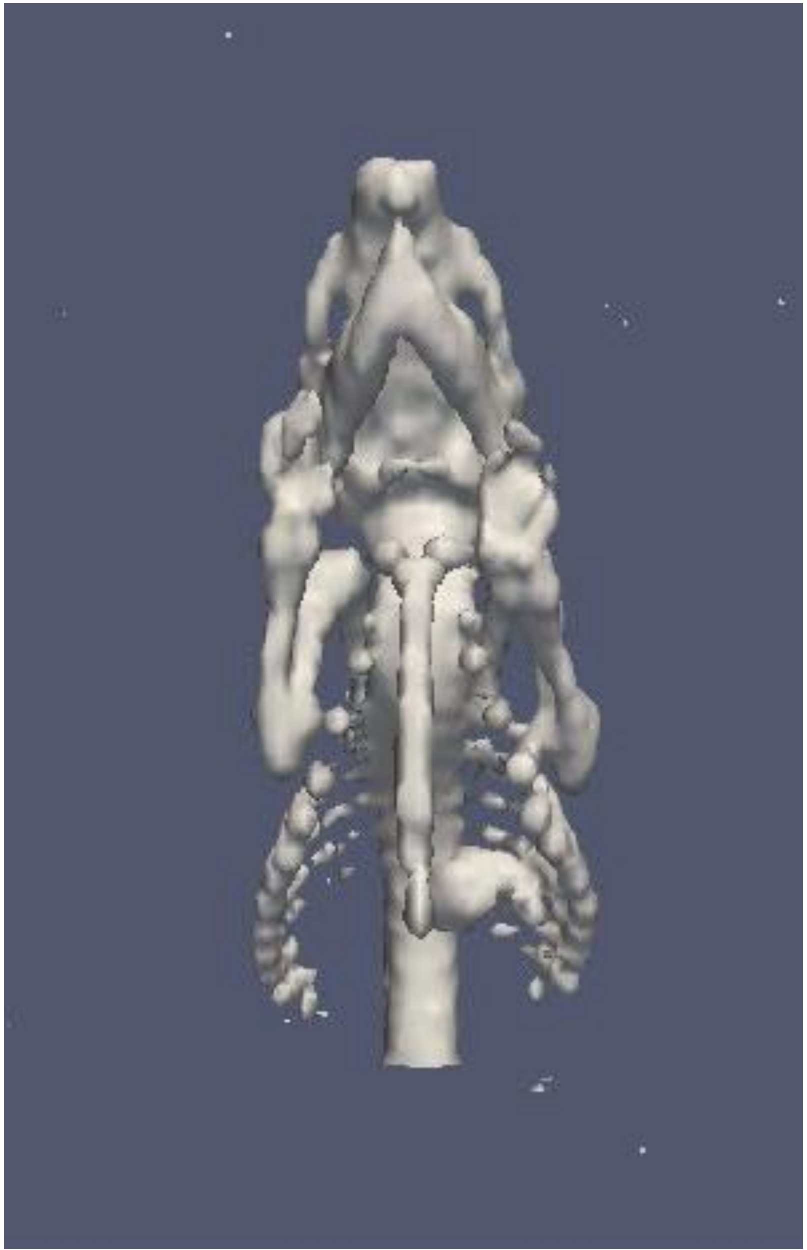
Rat, F-18 - Details of the skeleton



CT image from Explore Vista
(not enough activity left for the PET acquisition with Explore Vista)

Rendered F-18
image of rat

(click to animate)

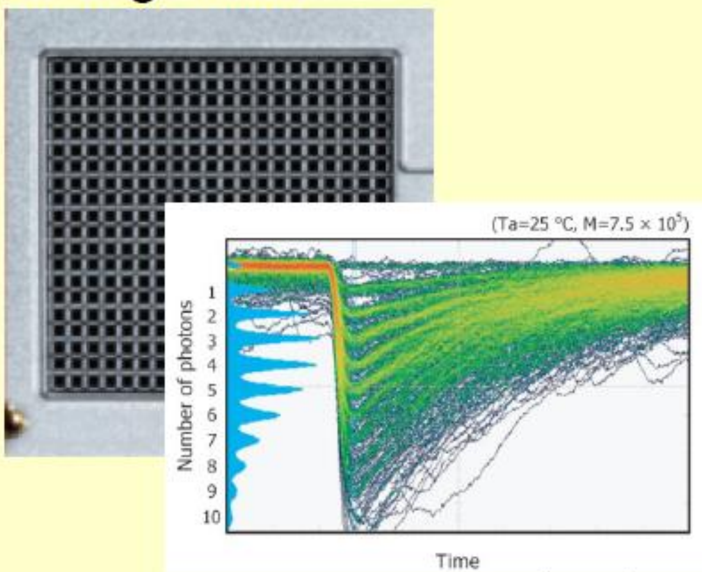


And finally … Digital SiPM

With the GM-APD being a binary device…
one could also conceive a fully digital SiPM, where the output is just the digital count of the number of fired cells.

Philips

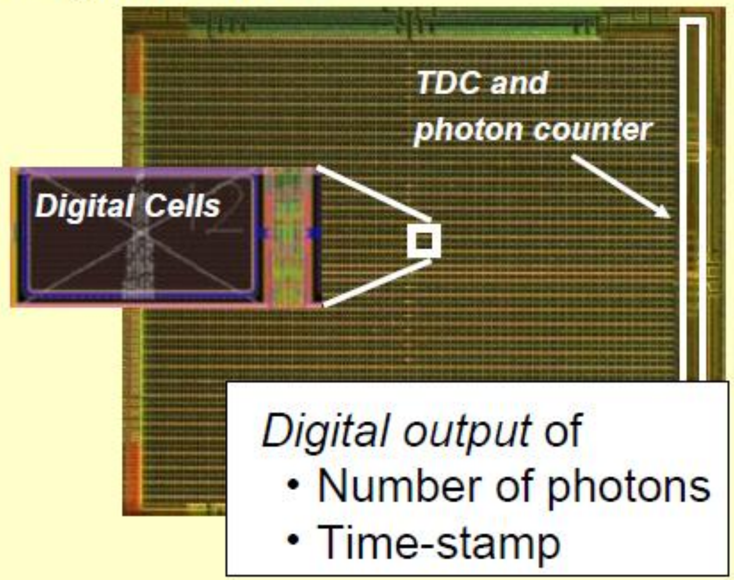
Analog SiPM



www.hamamatsu.com

- Cells connected to common readout
- Analog sum of charge pulses
- Analog output signal

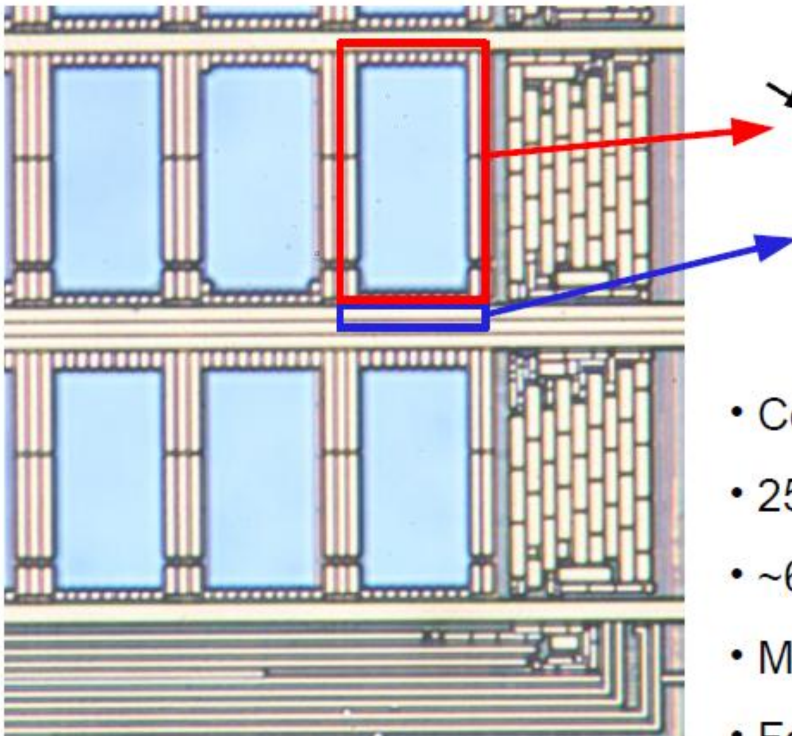
Digital SiPM

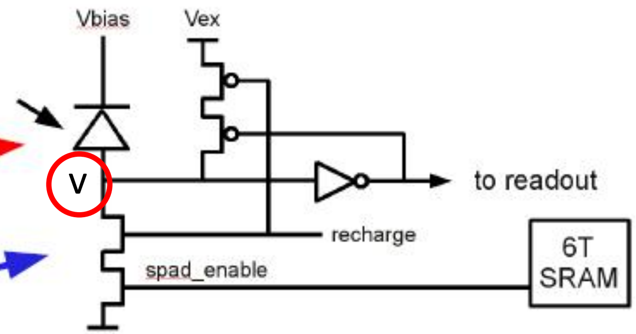


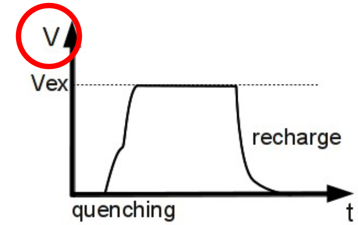
- Each diode is a digital switch
- Digital sum of detected photons
- Digital data output

Digital SiPM

1 digital cell (area depends on type, e.g. 30 x 50 μm)







- Cell electronics area: 120μm²
- 25 transistors including 6T SRAM
- ~6% of total cell area
- Modified 0.18μm 5M CMOS
- Foundry: NXP Nijmegen

Philips

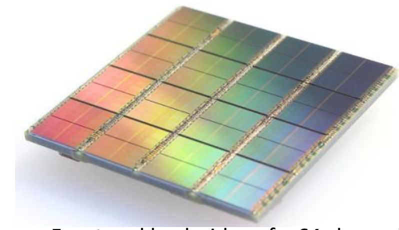
Different from analog SiPMs: Upon the detection of a photon, the avalanche is actively quenched using a dedicated transistor, and a different transistor is used to quickly recharge the diode back to its sensitive state.

Digital SiPM

Compared to the analog technology, the digital one (so far only offered by Philips) has a number of

advantages

+ Integration of bias supply, amp, TDC, counter…
+ Fast active quenching → no afterpulses
+ Possibility to de-activate noisy cells → potentially lower dark noise
+ Reduced sensitivity to voltage and temperature variations
+ Compactness
+ Possibility to add local intelligence



Front and back sides of a 64 channel digital tile (DPC6400-22-44) or (DPC3200-22-44)

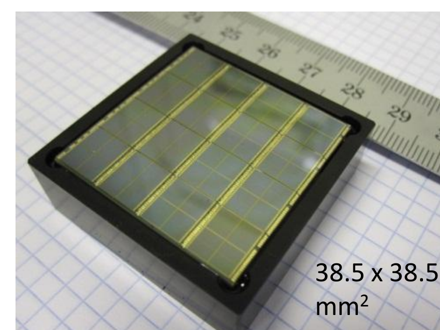
Philips

… **problems shared with analog**

○ High dark noise (a discharging cell doesn't know whether it is digital or analog)
○ Signal saturation (limited number of cells)

… and also has some **drawbacks**

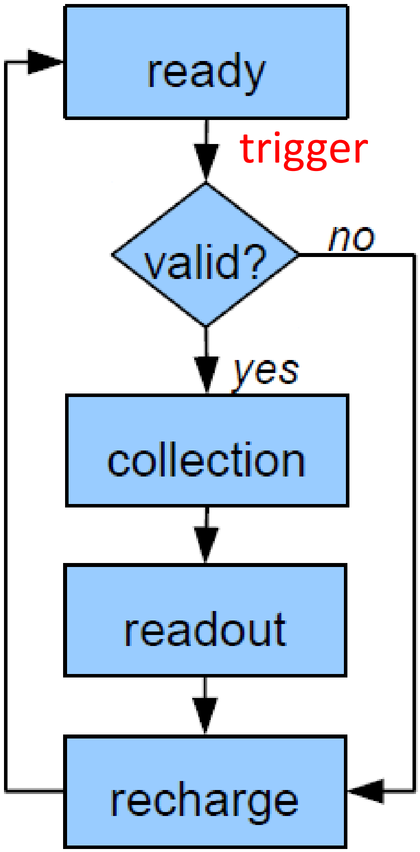
- The local electronics is a source of heat → cooling advisable
- The readout functionality is designed into the sensor. In case of mismatch with the needs, relatively expensive modifications of the sensor/FPGA may be required.



Packaged module, as delivered to clients (DPC3400-22-44). Includes a 100 μm thick protective glass layer.

Digital SiPM – State Machine

Philips



5ns – 80ns

0ns – 20µs

680ns

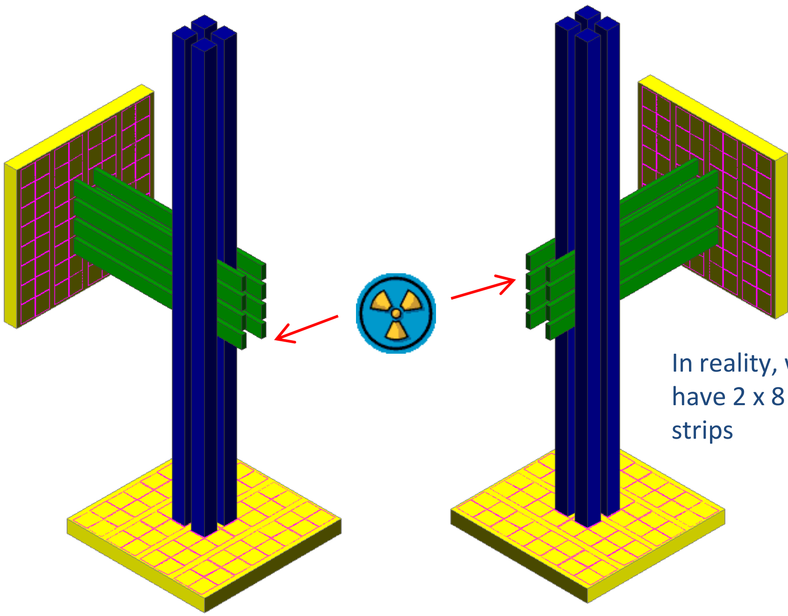
10ns – 40ns

- *200MHz (5ns) system clock*
- *Variable light collection time up to 20µs*
- *20ns min. dark count recovery*
- *dark counts => sensor dead-time*
- *data output parallel to the acquisition of the next event (no dead time)*
- *Trigger at 1, ≥2, ≥3 and ≥4 photons*
- *Validate at ≥4 ... ≥64 photons (possible to bypass event validation completely)*

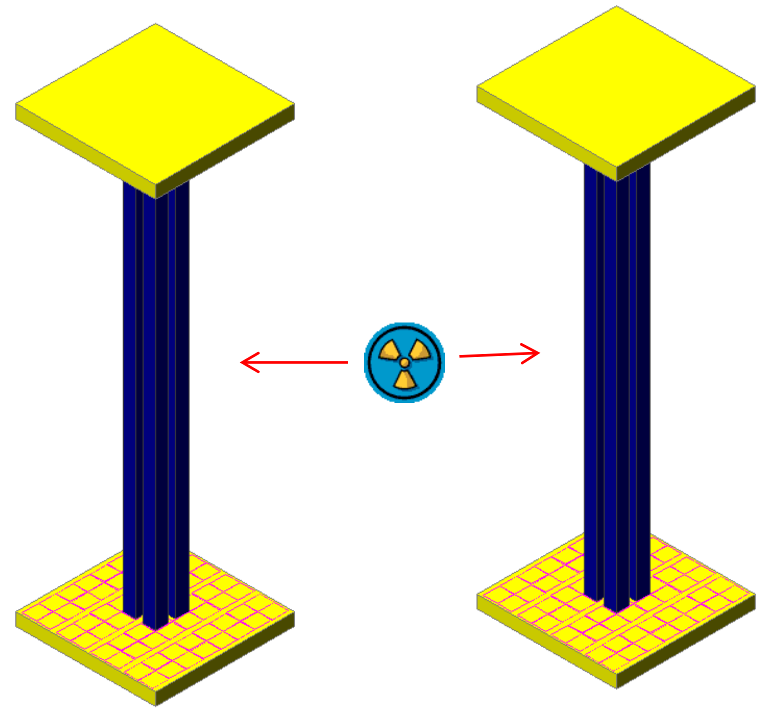
Every non-validated trigger leads in principle to the recharging of all cells. Without cooling, the device can loose efficiency/availability.

$\Sigma < 1\ \mu s$ Expect readout rate up to O(1 MHz)

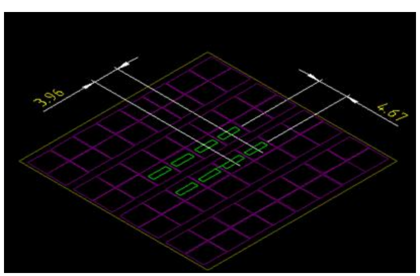
We got fascinated by the idea of a digital AX-PET module, mainly in view of a possible TOF-PET extension.

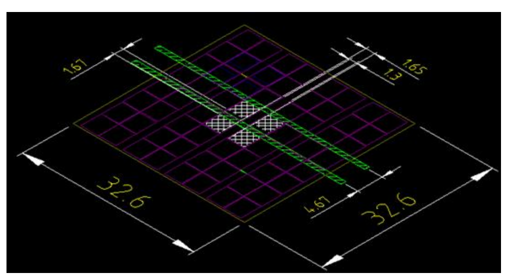


In reality, we have 2 x 8 WLS strips



Luckily, crystals and WLS would just fit (without big compromises) on the existing Philips 64 channel DSiPM modules DLS-3200 (6400)

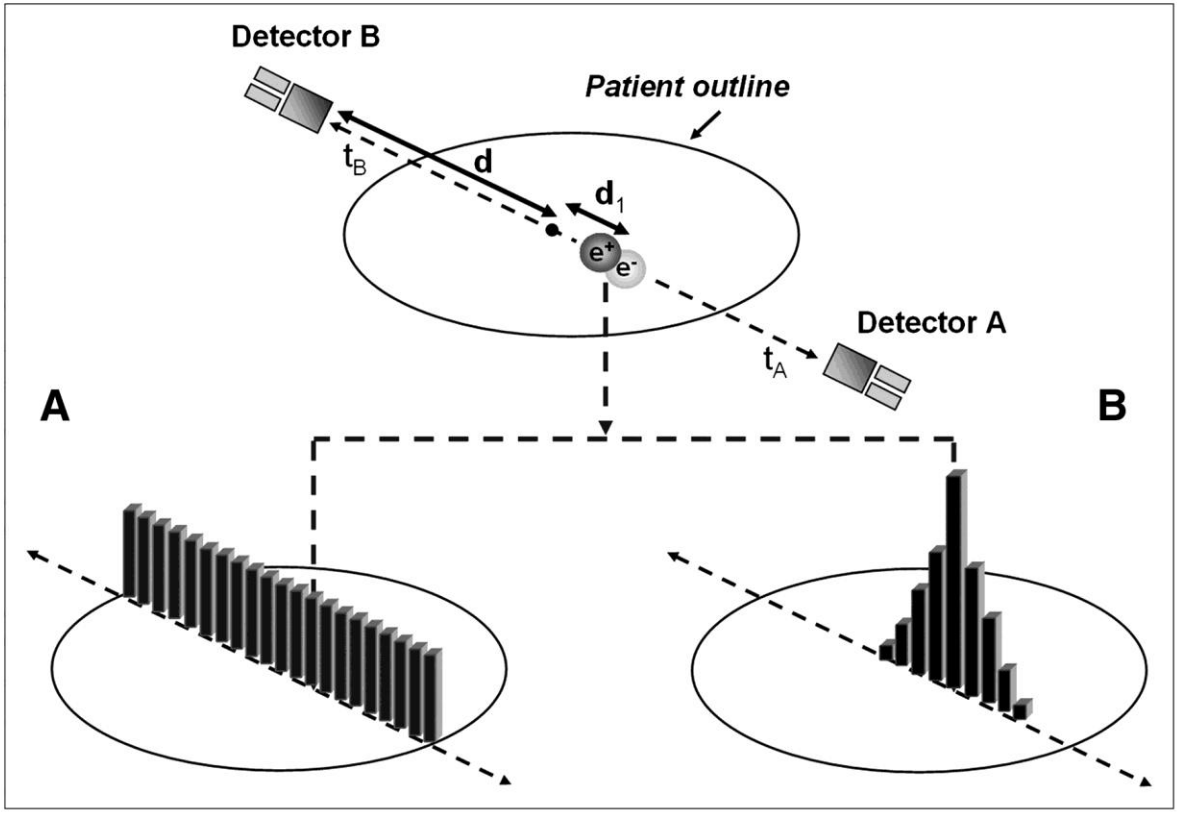




Aim for dual sided readout of long crystals to get rid of propagation delays

Principle of TOF-PET

Idea: use time information to localize annihilation along the line of response. LOR does no longer contribute uniformly to all pixels but according to the measured time and its resolution.
Images contain less noise and show better contrast (particularly noticeable for large scanners and big patients)



$\Delta t = 2 \cdot \Delta x / c$

→ Need δt = 66 ps
for δx = 1 cm

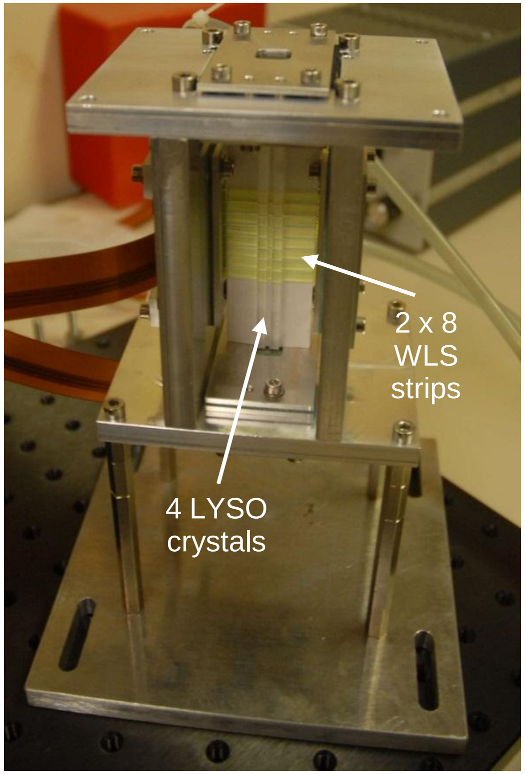
Gain of a TOF system over a non-TOF system is

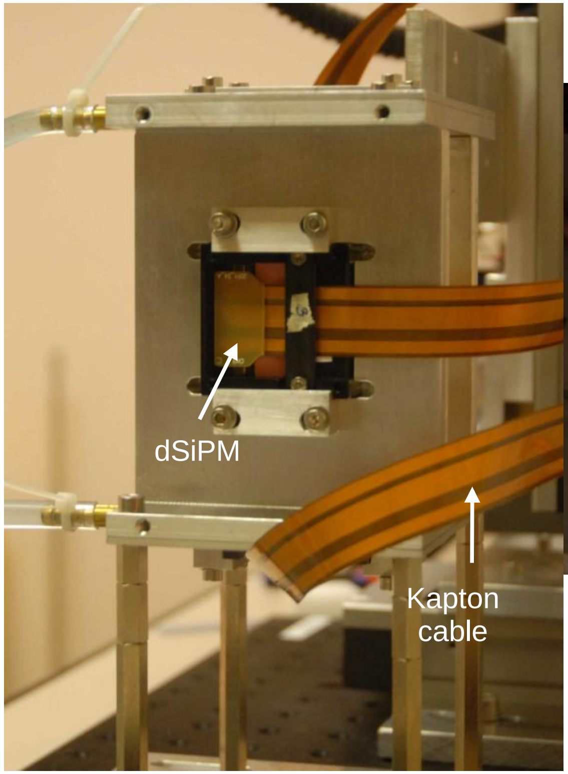
$G = 2D / c \cdot \delta t$

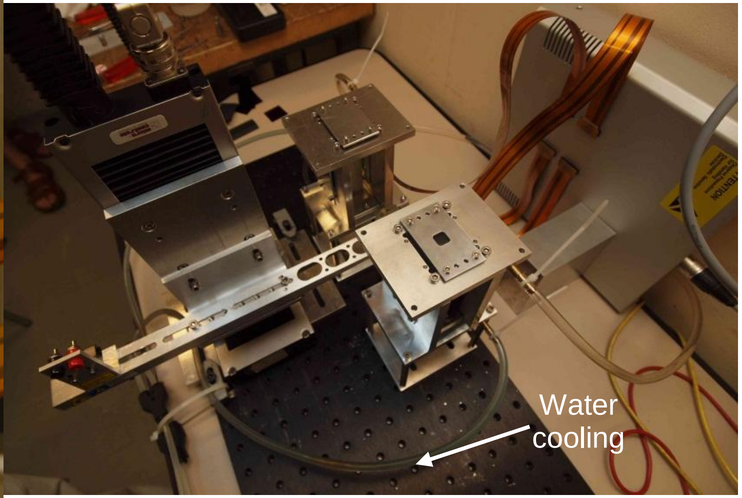
D = dimension of object being imaged.

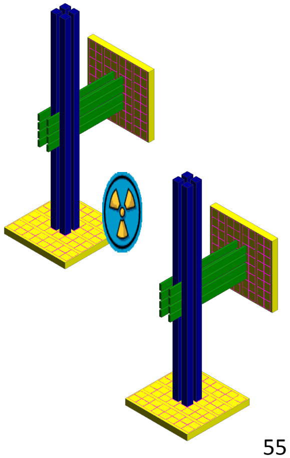
Verification of normal AX-PET characteristics

Two modules setup. Single sided readout plus WLS strips.







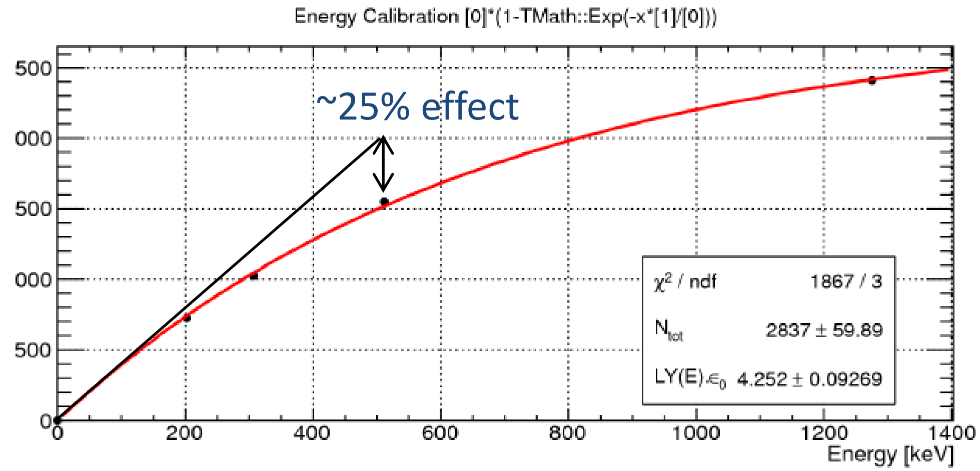


Energy calibration and resolution

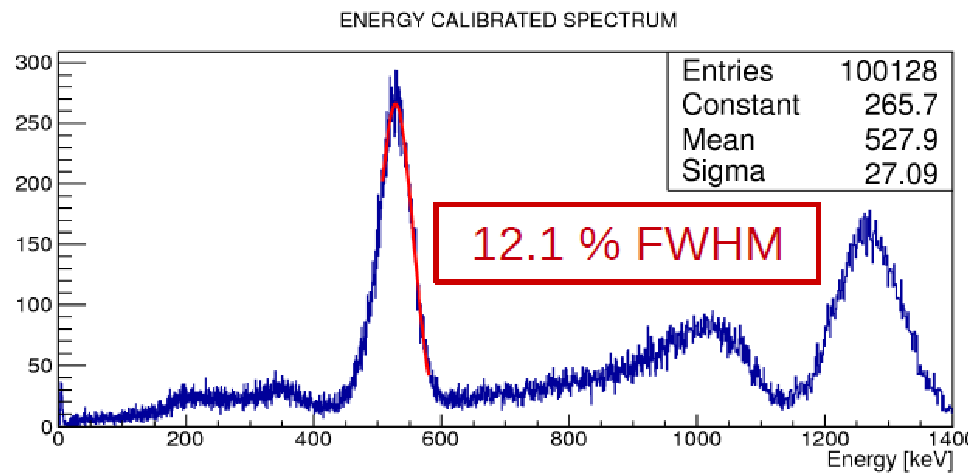
(single sided readout)

Saturation effect is much larger:

- More detected p.e.
- Fewer pixels (3200 vs 3600)
- Recharging only after readout!

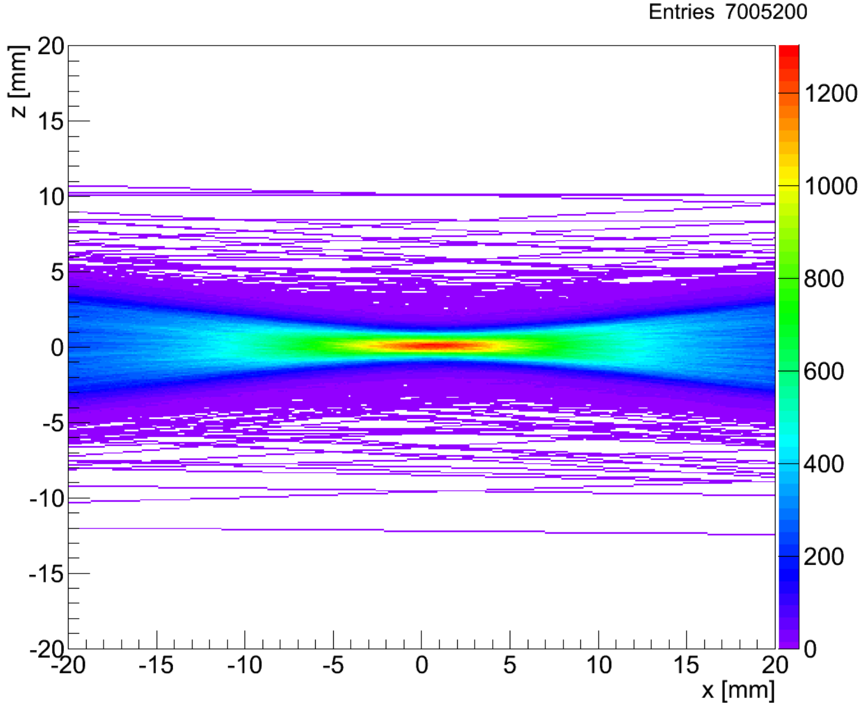


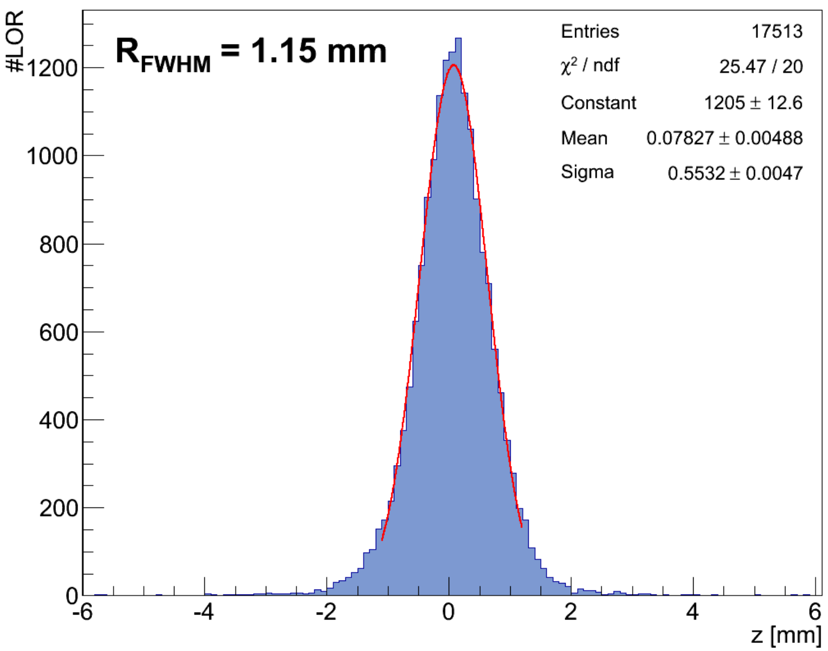
In the end we get a comparable Energy resolution.
(analog AX-PET was 11.7 %).



Confocal reconstruction in axial direction (WLS array)

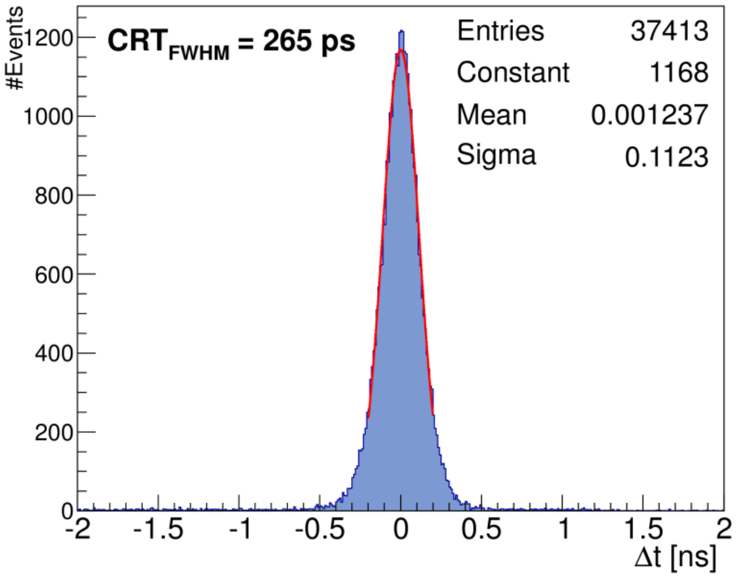
^{22}Na source of 250 microns diameter, 150 mm between the two modules



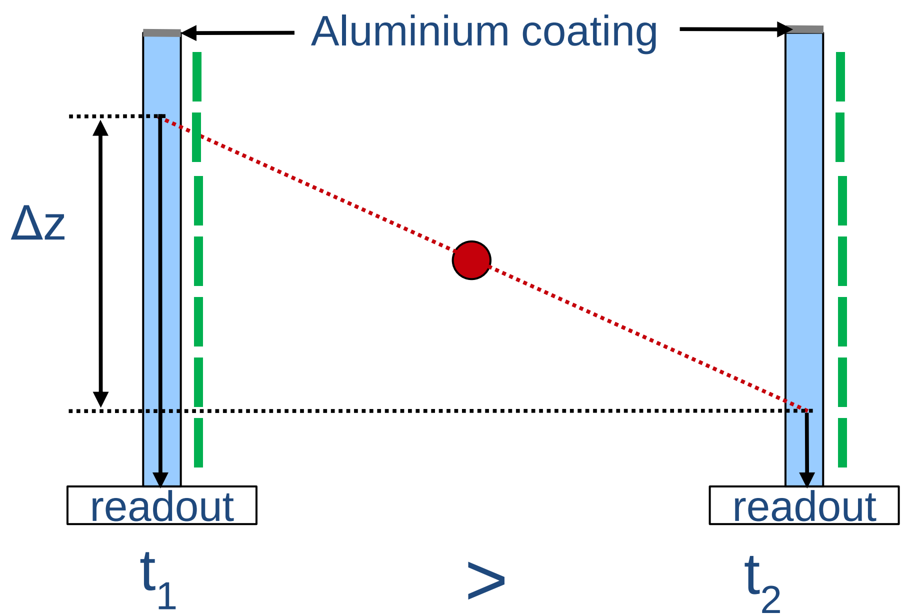


Slightly better than with the analog AX-PET set-up (1.48 mm)

Coincidence time resolution
(single sided readout)

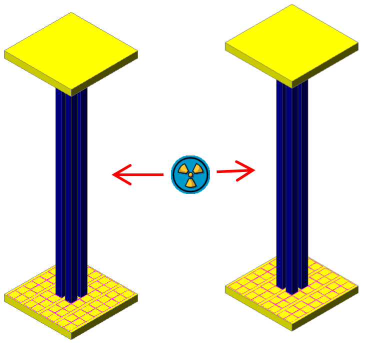


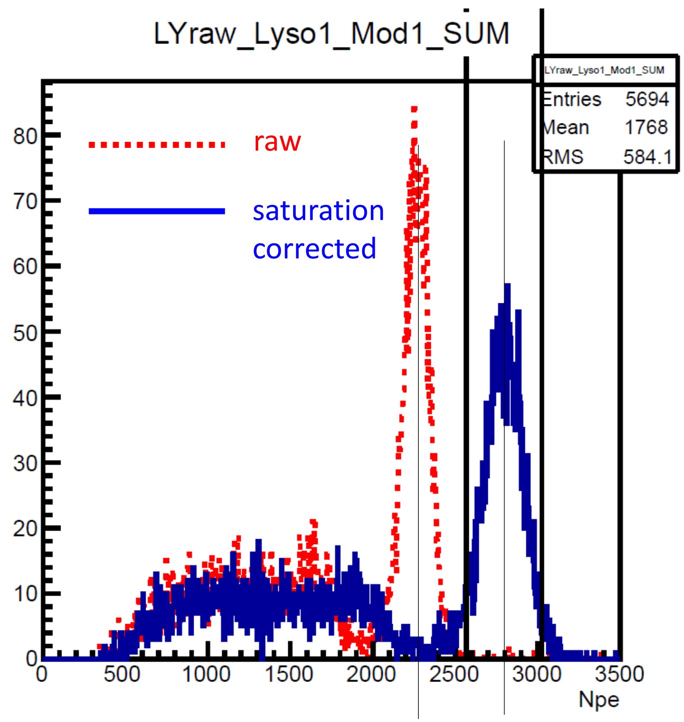
This is only possible after correcting for light paths using WLS information



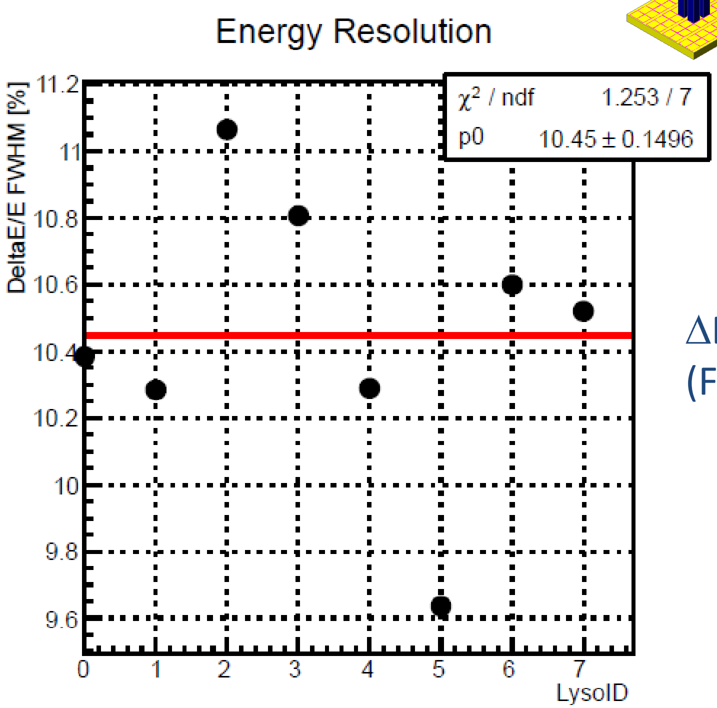
Energy calibration and resolution

(Double sided readout)





2250 → 2800
(25%)



ΔE = 10.45%
(FWHM)

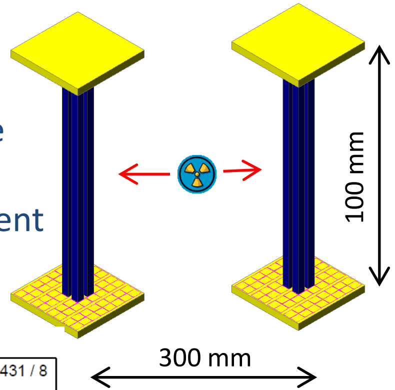
Coincidence time resolution by mean timing

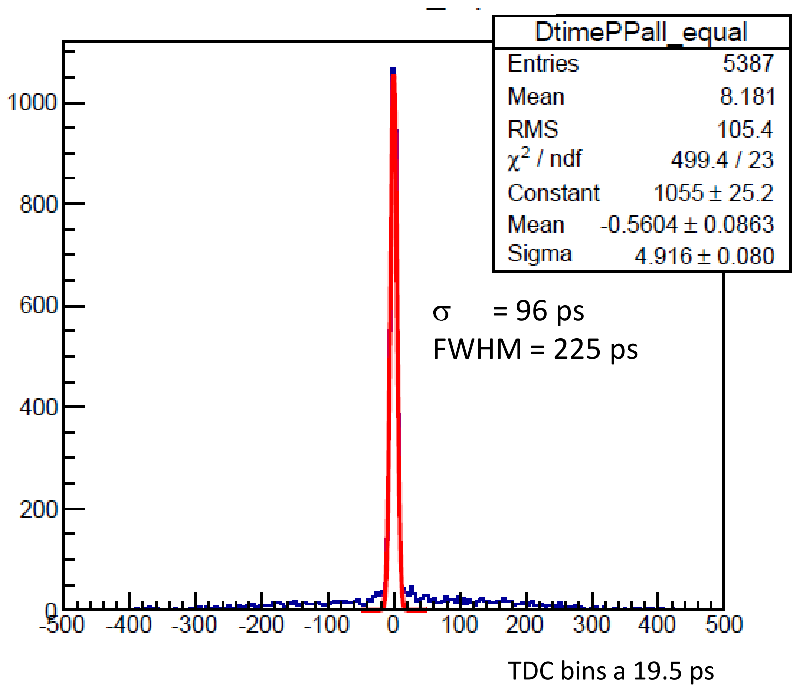
(Double sided readout)

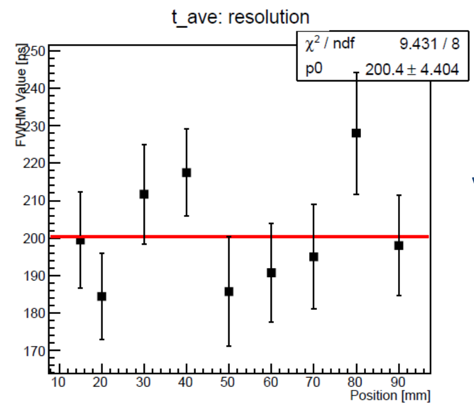
$$\Delta t = \langle t_{mod1} \rangle - \langle t_{mod2} \rangle = \frac{t_{1,up} - t_{1,down}}{2} - \frac{t_{2,up} - t_{2,down}}{2}$$

Small differences in timing of pixels (offsets) can be eliminated.

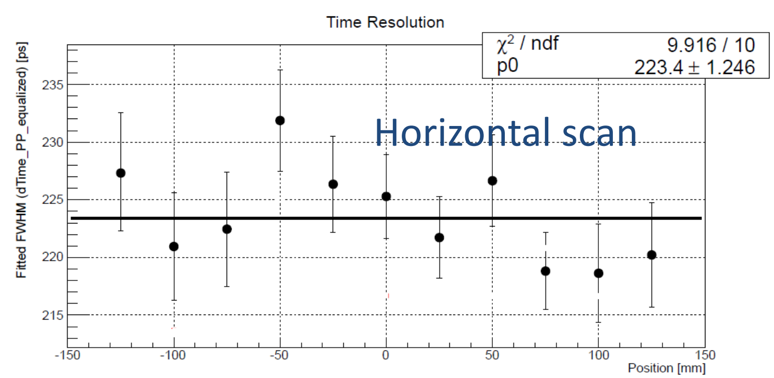
Mean Timing corrects automatically for light propagation times in the crystals. The resolution becomes fully independent of the source position in the FOV.







Vertical scan (1 module + small tagger)



Brief look at applications in particle physics

The SiPM technology wasn't mature enough when the choices were made for LHC (>10 years ago).
But we start to see SiPMs in R&D for the LHC upgrade.

The obvious applications are readout of scintillators and scintillating fibres.
Also demonstrated: Detection of Cherenkov light (even with digital SiPM).

- First major project was T2K (scintillator tracker + ECAL). ~65k SiPMs.
- The CALICE collaboration builds HCAL and ECAL prototypes based on small scint. plastic tiles (ILC). ~8k SiPMs.
- CMS is working on an upgrade of HCAL barrel (SiPMs replacing HPDs).
- LHCb is working on a study of SciFi tracker to replace Si strips and gas straw tubes.
  ~ 300k SiPMs
- CLIC detector study for a W/scint ECAL. Many 100k SiPMs.

SiPM in the ILC HCAL and ECAL (R&D)

High granularity hadronic calorimeter optimised for the Particle Flow measurement of multi-jets final state at the ILC

$e^+ e^-$ collider (1 TeV)

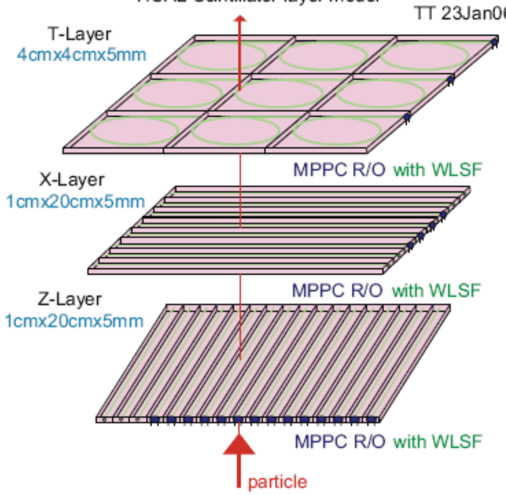
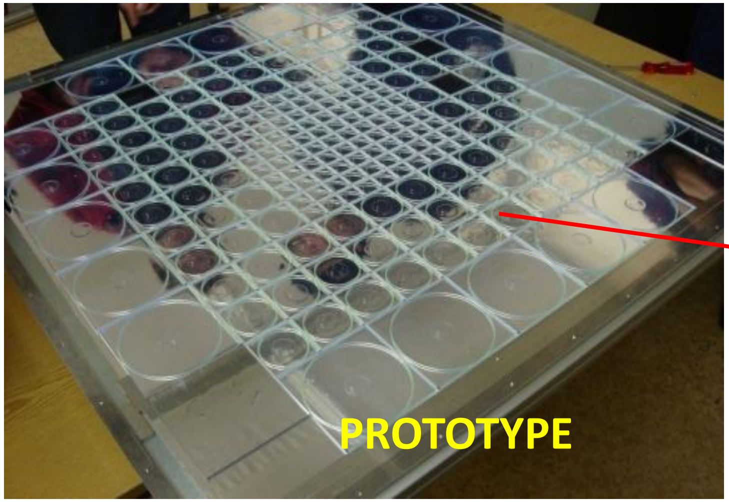


Fig. 6. A possible scintillator strip/tile sequence of the analog HCAL.

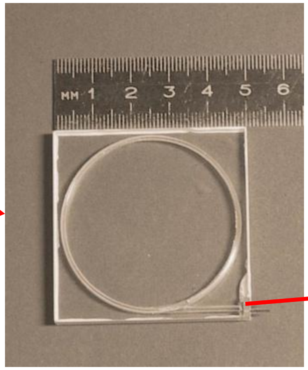
photo-detector requirements:

- insensitive to magnetic field (~ 5T)
- coupling with a scintillator (blue emission)
- stability (T-control or T-monitoring)
- rel. large dynamic range
- low cost

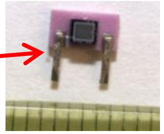
CALICE tests with MePHI/PULSAR, HAMAMATSU, KETEK SiPMs



216 tiles/layer (38 layers in total) ~8000 channels



3 x 3 cm² plastic scintillator tile with embedded WLS fiber + SiPM



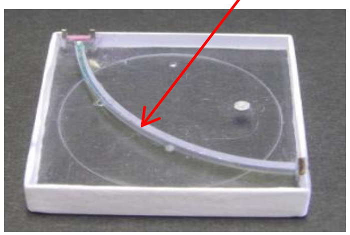
SiPM
1 mm²

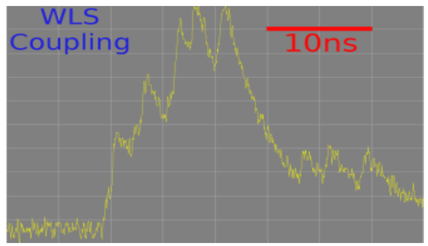
Readout of SiPMs by the SPIROC ASIC (LAL)

SiPM in the ILC HCAL (R&D)

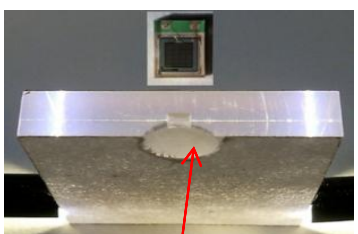
Different SiPM couplings to scintillator tile

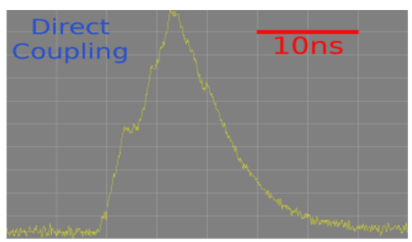
with WLS fiber (for MEPHI SiPM)





direct coupling (for MPPC)



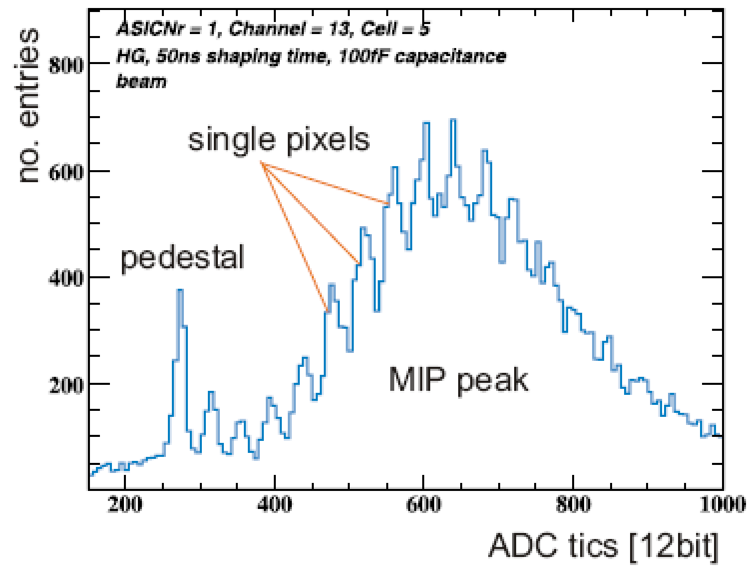


hole ("dimple") for more uniform light yield close to photodetector

Calibration & monitoring

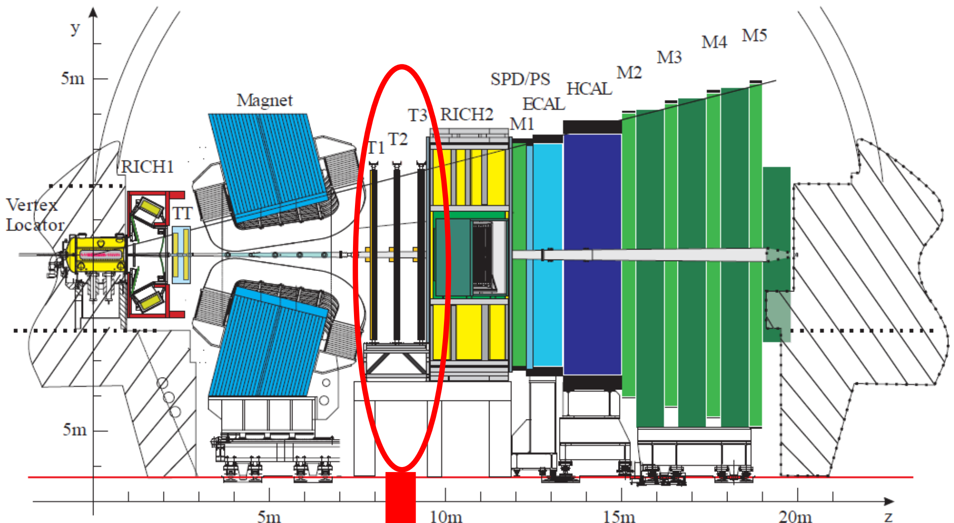
- test beam (absolute)
- LED monitoring system (stability)

DESY 3 GeV electron testbeam

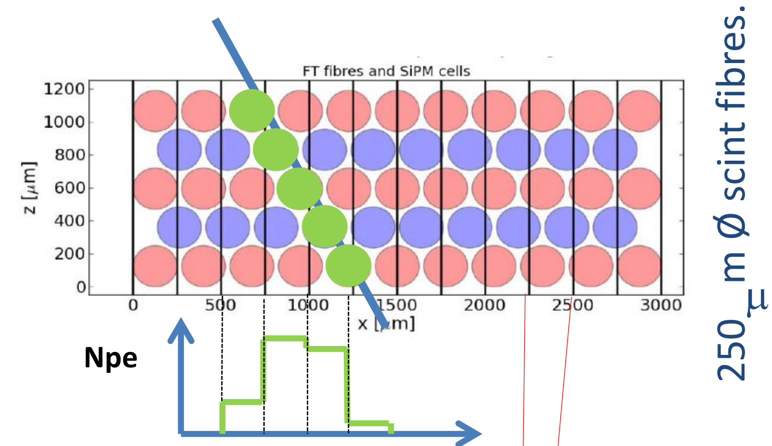


- Channel energy calibration by ~3 GeV electrons (MIPs).
- Multi-photon peaks allow simple absolute calibration.

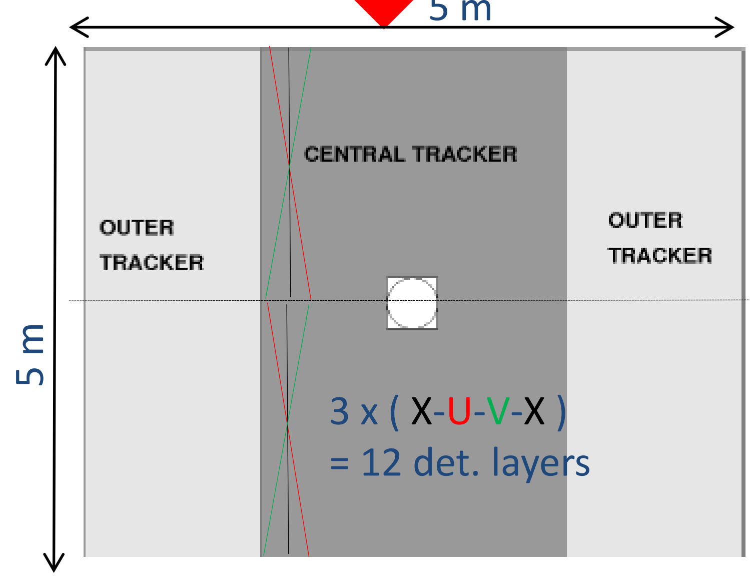
R&D for LHCb SciFi tracker



1 det. layer = 5 fibre layers



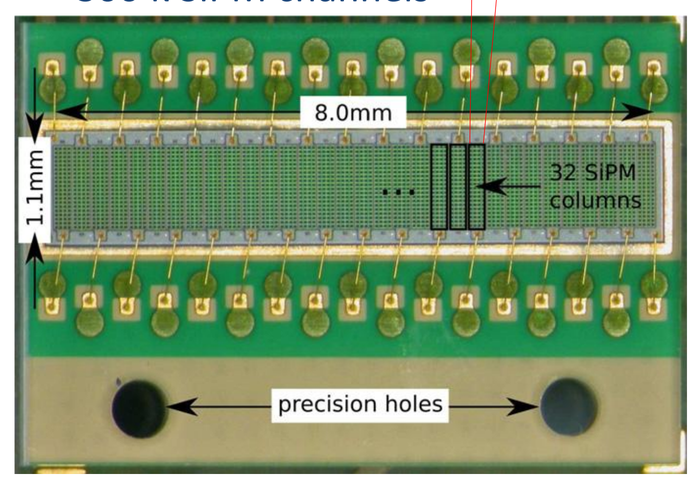
~20 pe, σ_x ~ 50 μm



Detector concept follows largely the work of the Aachen group

(S. Schael et al.)
NIM A 622 (2010) 542–554

~ 300 k SiPM channels



Main challenges:

- **Rad hardness of SiPM** (10 years ~ $6 \cdot 10^{11}$ n/cm^2)
- (Rad hardness of scint. Fibres (10 years ~ 30 kGy))

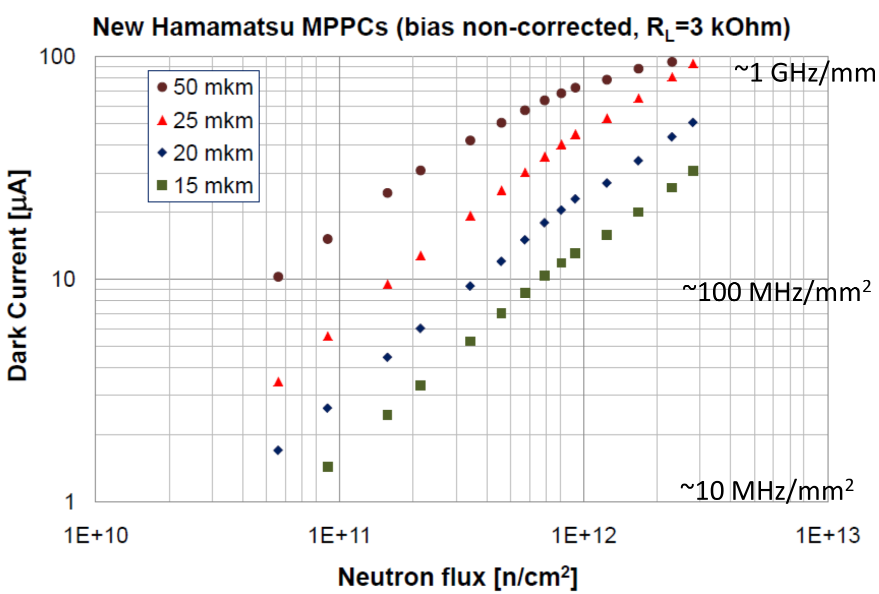
Main observation: dark current / dark rate increase linearly with the NIEL

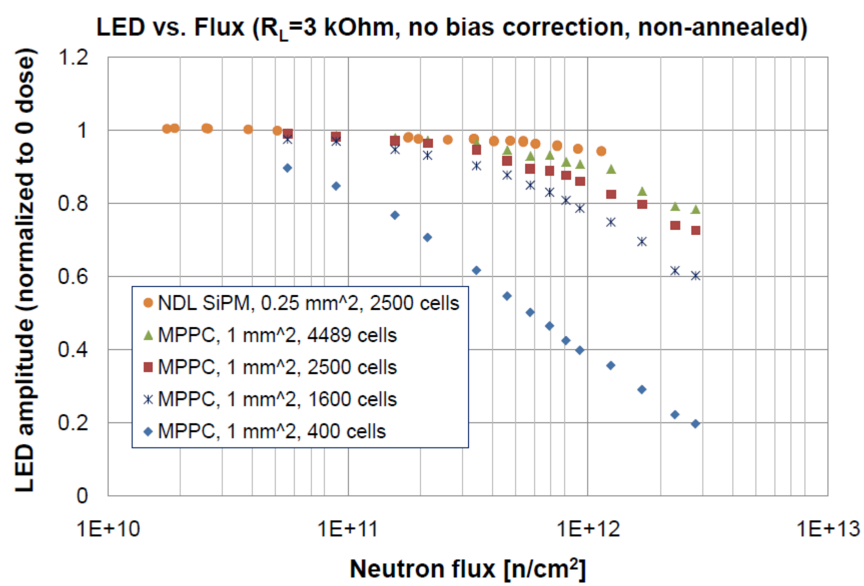
In analogy to a standard pn detector…

$$I_d \sim \alpha \cdot \Phi_{(1\ \mathrm{MeV\ n\ eq.})} \cdot V_{eff} \cdot G \quad (\alpha = 3 \cdot 10^{17}\ \mathrm{A \cdot cm})$$

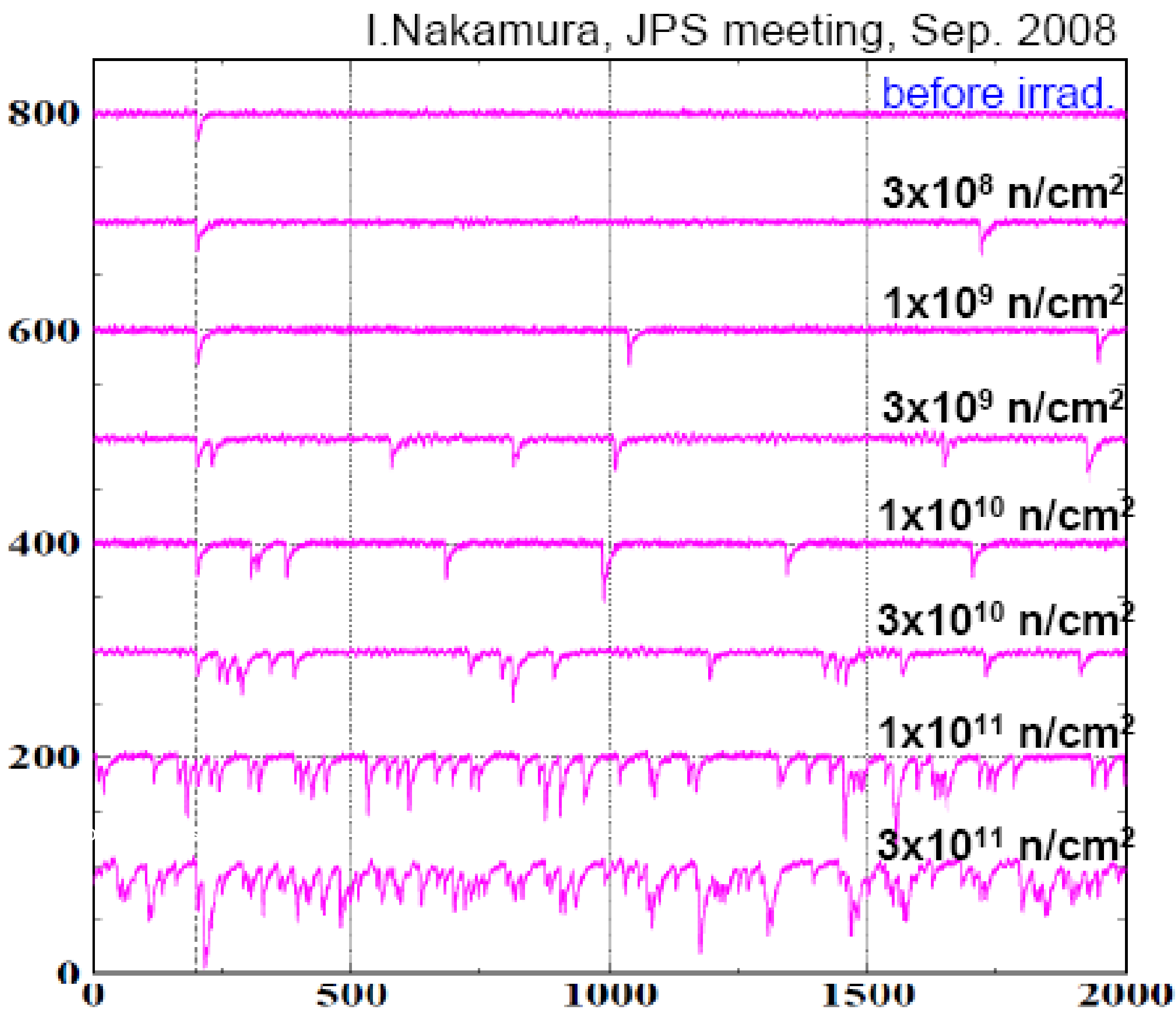
But also gain starts to drop

Devices with smaller cell size and fast cell recovery survive longer



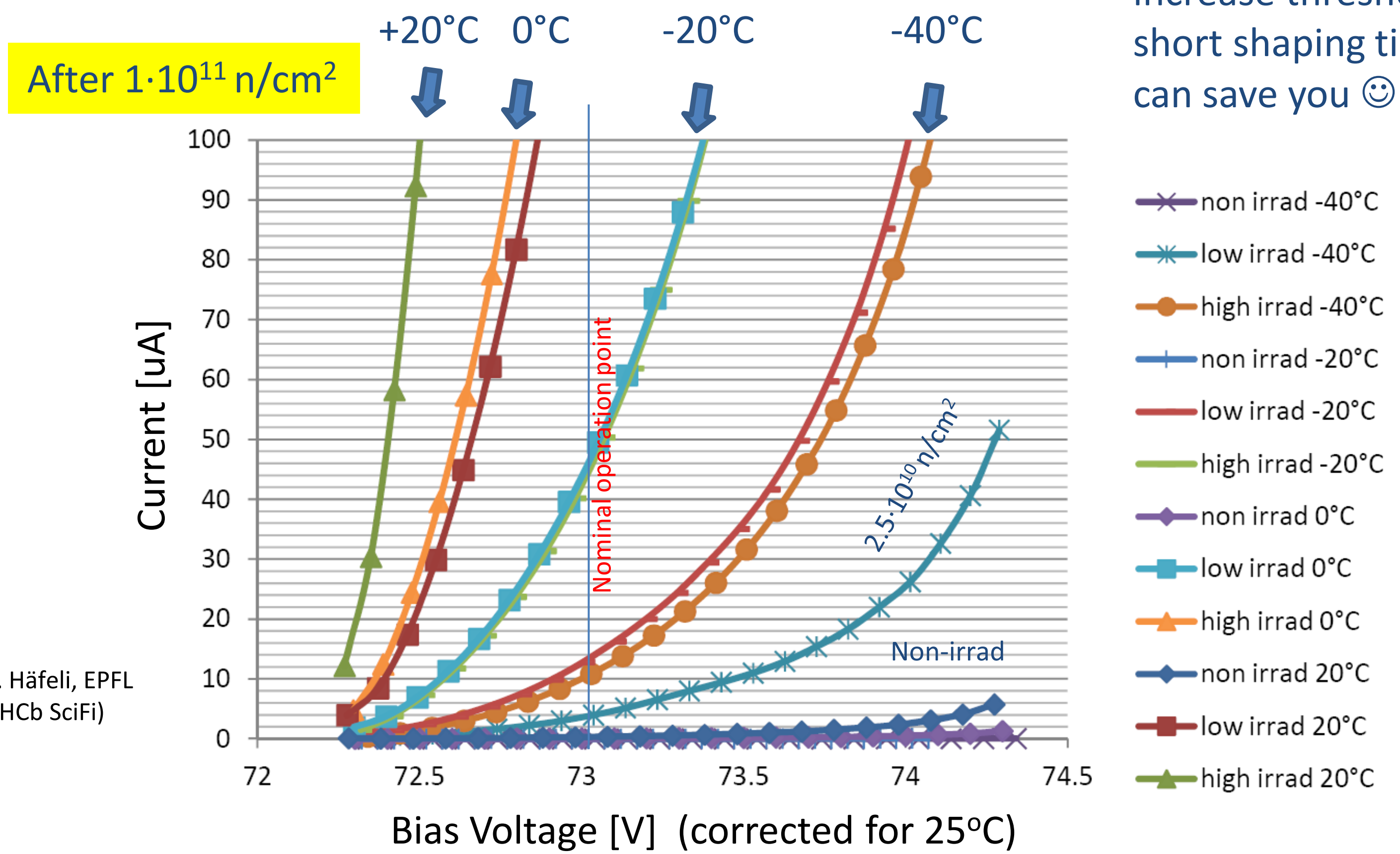


I. Musienko, SiPM workshop, CERN 2011.


I.Nakamura, JPS meeting, Sep. 2008
before irrad.
3x10^8 n/cm^2
1x10^9 n/cm^2
3x10^9 n/cm^2
1x10^10 n/cm^2
3x10^10 n/cm^2
1x10^11 n/cm^2
3x10^11 n/cm^2
800
600
400
200
0
0
500
1000
1500
2000

time (ns)

Cooling helps a lot. Every 8K reduce the dark noise by a factor 2



G. Häfeli, EPFL
(LHCb SciFi)

But also reducing overvoltage, increase threshold, clustering, short shaping time, … can save you ☺

Dark current (Hamamatsu SiPM module of 128 ch à 0.25 × 1.1 mm^2 ~ 35 mm^2)

Summary / Conclusions

SiPMs are photodetectors with great properties (sensitivity, gain, compactness, simplicity) and a few limitations (dark noise, rad. tolerance, dynamic range, rel. high cost).

In the past decade they have stimulated huge interest (by research teams, industry) and their properties improved a lot. They enabled new applications and will continue to do so.

Digital SiPM technology is a consequent next step. Their acceptance by the market depends on many other aspects.

AX-PET is an example of a detector concept which would be very difficult to implement without SiPMs. It overcomes the sensitivity-resolution dilemma of standard PET detectors. Its acceptance by the market …

Many thanks to my AX-PET colleagues Chiara Casella (ETH Zurich) and Matthieu Heller (CERN) for many plots and schematics used in this talk. Thanks also to Felix Sefkow for CALICE material.


Many thanks for
your interest !