The Silicon Photomultiplier for Application in PET

D.J. Herbert[1], S. Moehrs[1], N. D'Ascenzo[1], N. Belcari[1], A. Del Guerra[1],
V. Saveliev[2], M. Mandelkern[3]

[1] Dipartimento di Fisica & INFN, Università di Pisa, Pisa, Italy
[2] Obninsk State University of Nuclear Engineering, Obninsk, Russia
[3] UCI, California, USA

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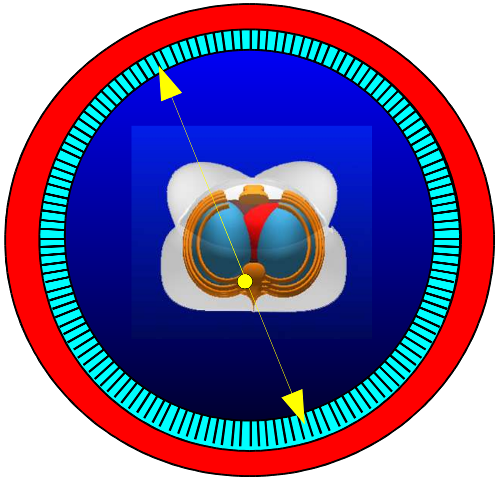
Positron Emission Tomography (PET)

- A β^+ emitting radiotracer is injected into a "patient"
- The radiotracer marks a specific function

  (e.g. glucose metabolism)

- The positron annihilates with an electron and produces a pair of opposite collinear 511keV photons
- A set of detectors surrounding the "patient" detects the pair of photons in time coincidence
- 3-D reconstruction gives the activity density within the body



- ***PET camera requirements – in general***

  - position information (~mm)
  - timing (~ns)
  - energy resolution (<20%)
  - all this with maximum possible sensitivity

Particular issues for prostate imaging

- The postion of the source is more or less known
- Hot areas in the region (bladder) giving a high singles rate
- Scattering in the body
- Specificity of current tracers is low

QUESTIONS TO BE ANSWERED

- Necessary resolution (<2mm??)
- Sensitivity – depends upon tracers
- Imaging area - the prostrate or also surrounding area?

High sensitivity, compact system with moderate spatial resolution

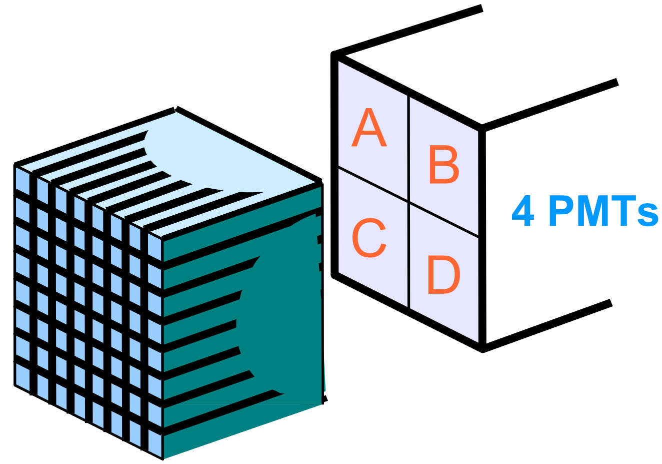
Making flexible new gamma camera systems

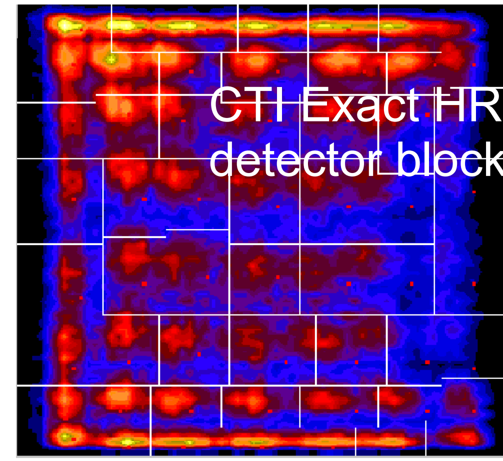
High-sensitivity, compact system with moderate spatial resolution

- **High sensitivity** – solid angle coverage
  -- stopping power of the scintillator
  -- thickness of scintillator
  -- dead time of system (Fast scintillator and readout)
- **Compact detector with low dead space**
- **Moderate resolution** – Pixelated or continuous?
- **COST?**

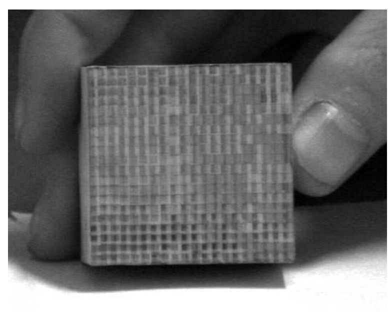
Detectors for PET

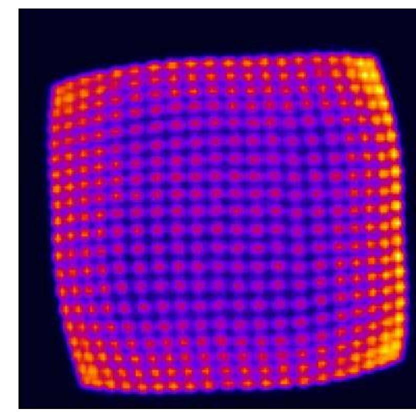
- Classic block detector (BGO then LSO) is still used but *limited in spatial resolution*





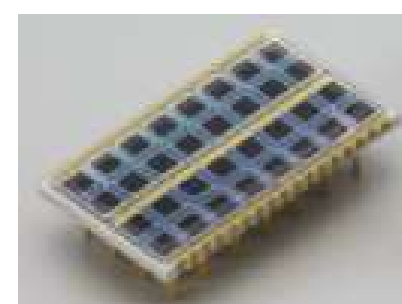
- For higher resolution, PSPMTs and pixelated scintillator arrays are used but they are of *limited size and expensive.*





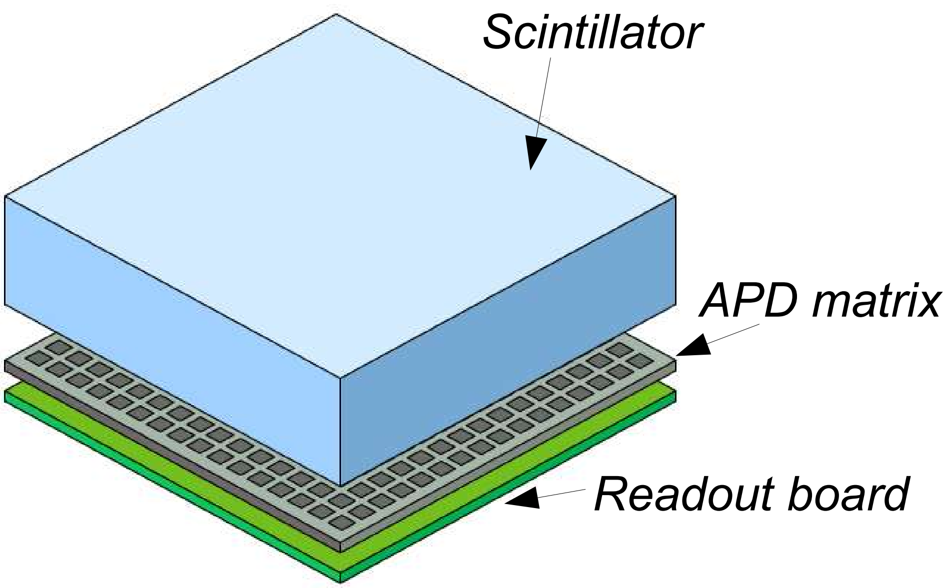
Arrays of semiconductor detectors- ie) APD matrices , offer many possibilities for creating compact, application specific cameras

- *limited gain*
- *cost?*

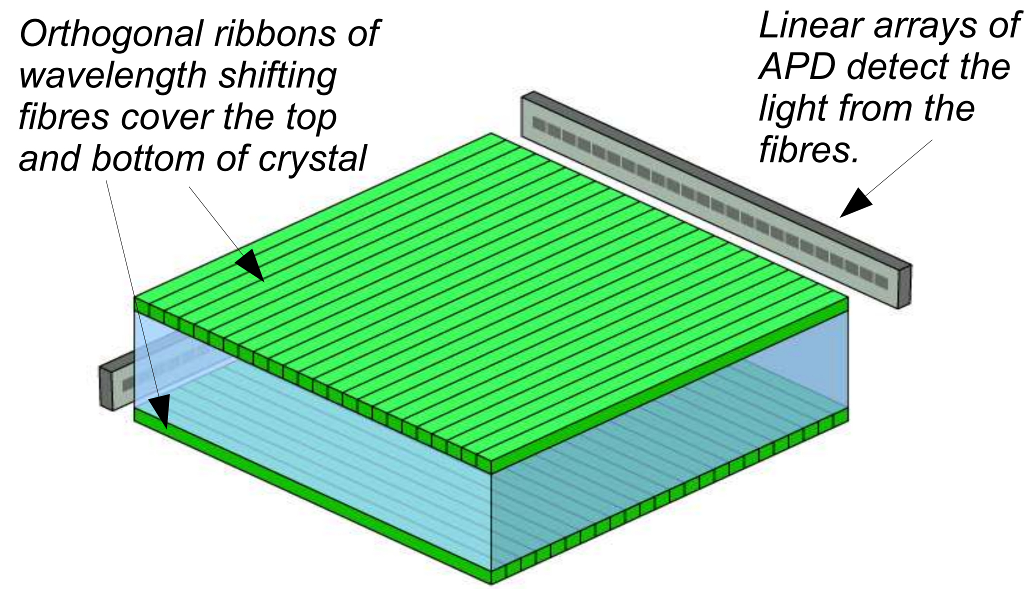


Potential configurations

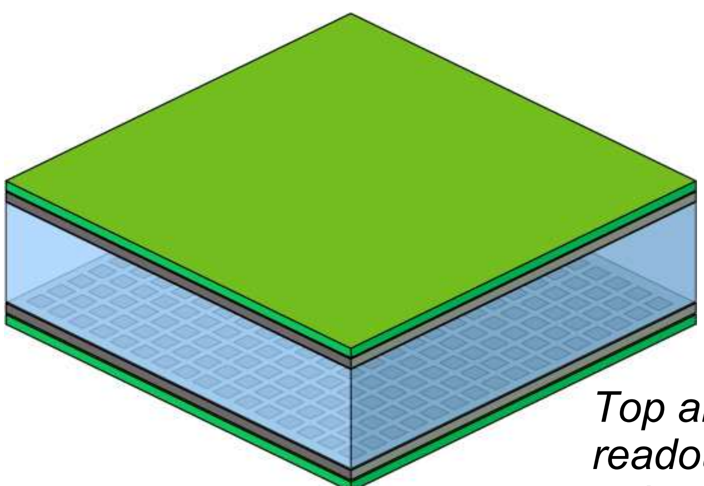
Anger camera



WLS Fibre Readout

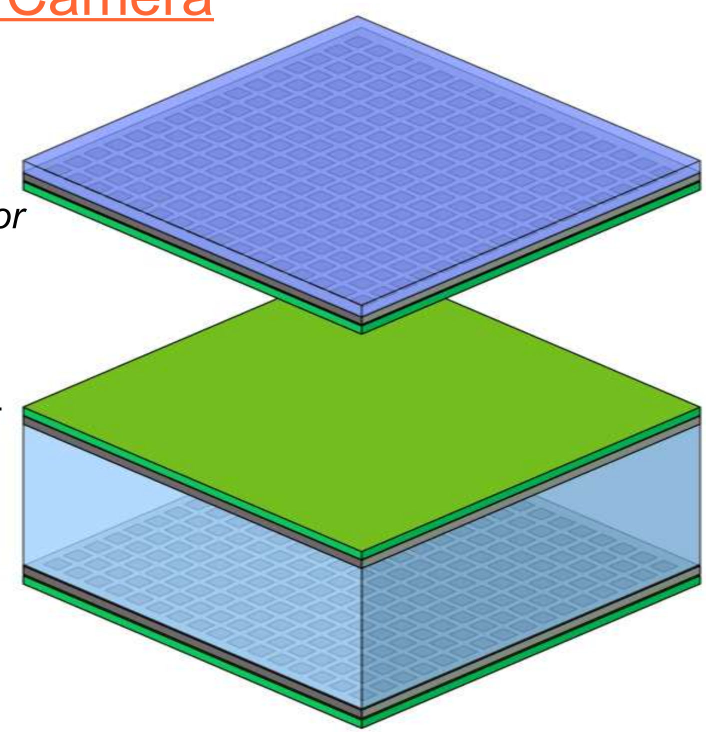


Double sided readout



Top and bottom readout can be used to have DOI.

Compton Camera

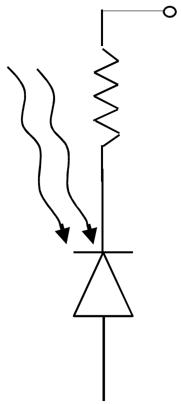


***Scatter detector** - thin, low Z scintillator with APD readout*

***Energy detector** – efficient scintillator with double sided readout*

A new detector – Silicon Photomultiplier (SiPM)

GM-APD + $R_{quenching}$



Output signal is independent of number of photoelectrons
(binary device)

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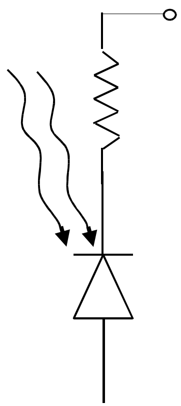

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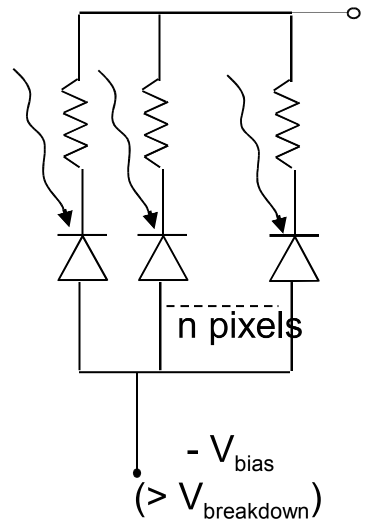

SiPM

- matrix of n pixels (~1000) in parallel
- each pixel: GM-APD + $R_{quenching}$

Output signal is
the sum of the signals
from all fired pixels
(analogue device)

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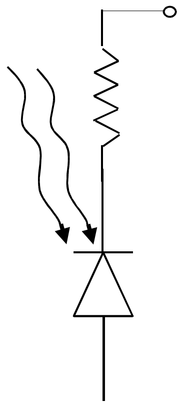

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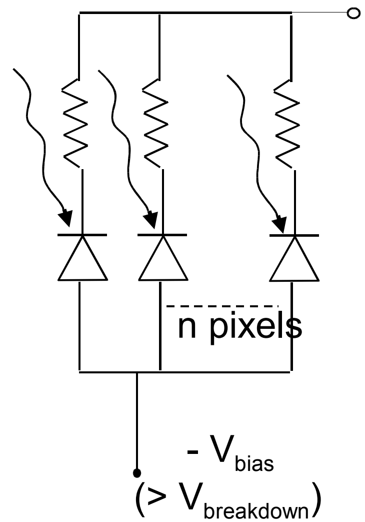

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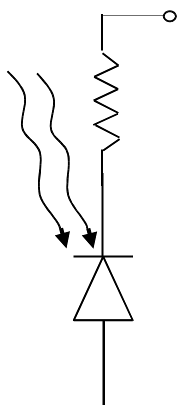

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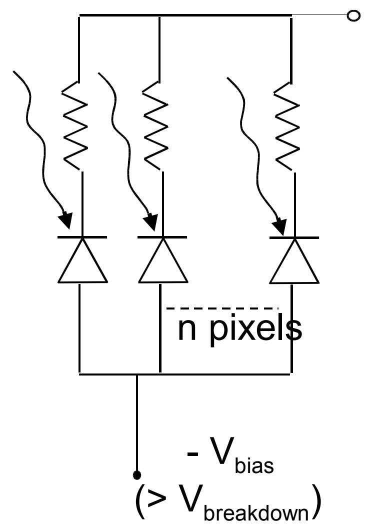

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Possible replacement for vacuum PMT?

- ✔ High internal gain (10^6) at low bias voltage (~50V)
- ✔ Excellent photon counting capability
- ✔ Fast response time (~1ns)
- ✔ Noise (dark counts) limited to photoelectron level
- ✔ Insensitivity to magnetic fields
- ✔ Compact and rugged

A new detector – Silicon Photomultiplier (SiPM)

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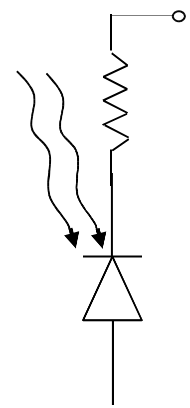

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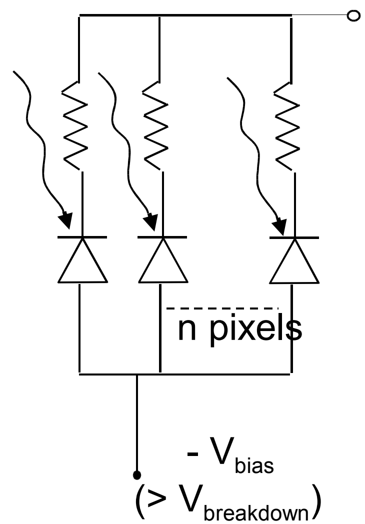

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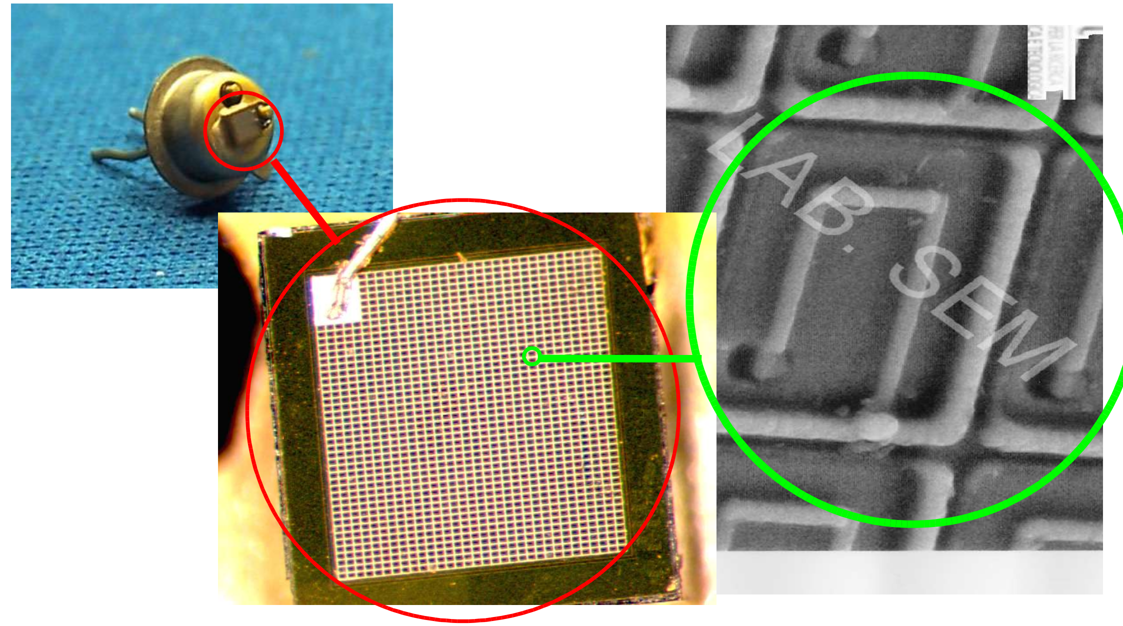
- ✗ Low detection efficiency (ave. 2.5% over LSO emission)
- ✗ Limited dynamic range (1000/mm^2)
- ✗ Not yet available in matrices

SiPM testing



- Samples from CPTA, Russia
- Bias ~ 50V
- Fast preamp + QDC

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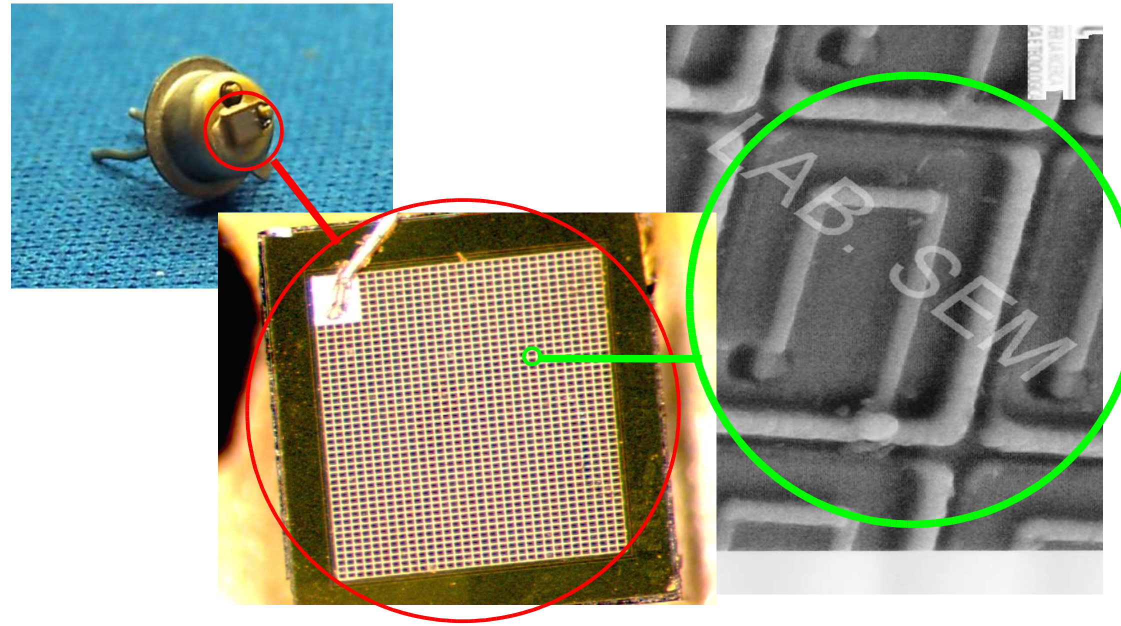

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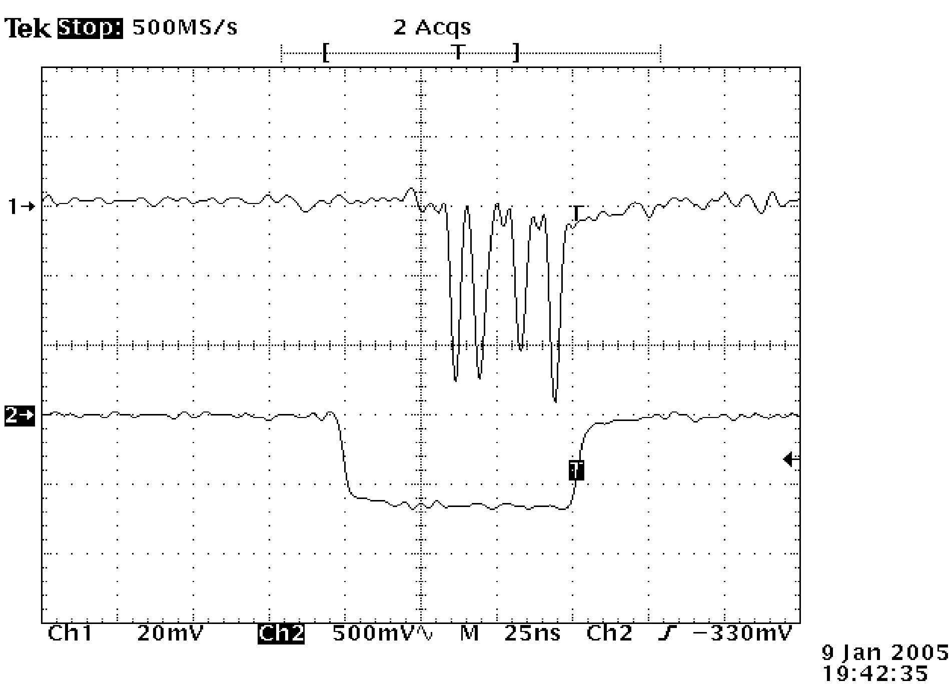

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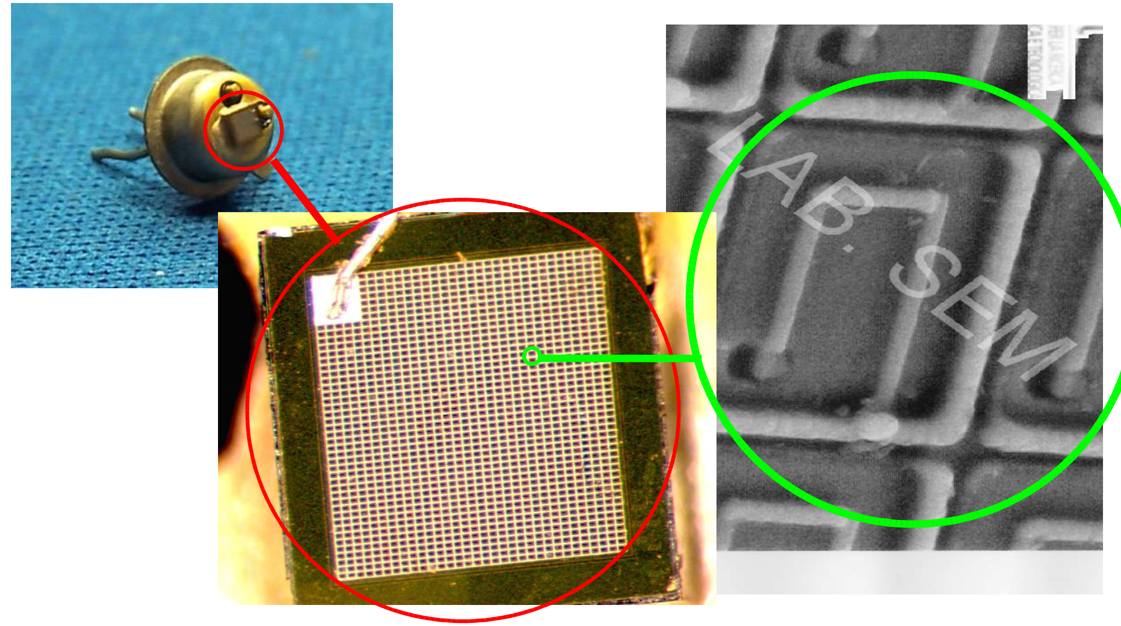

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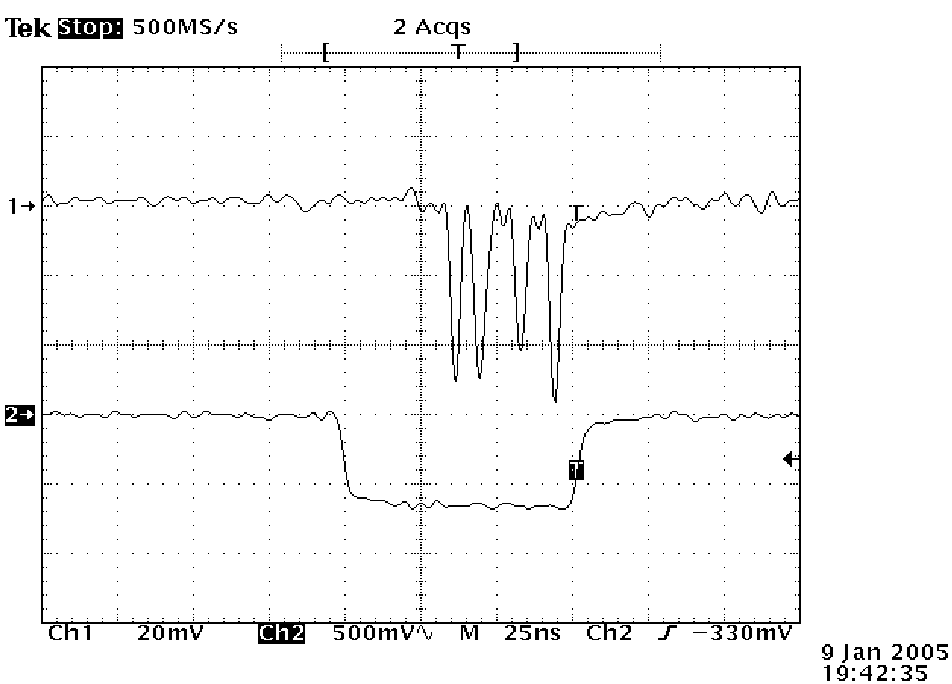
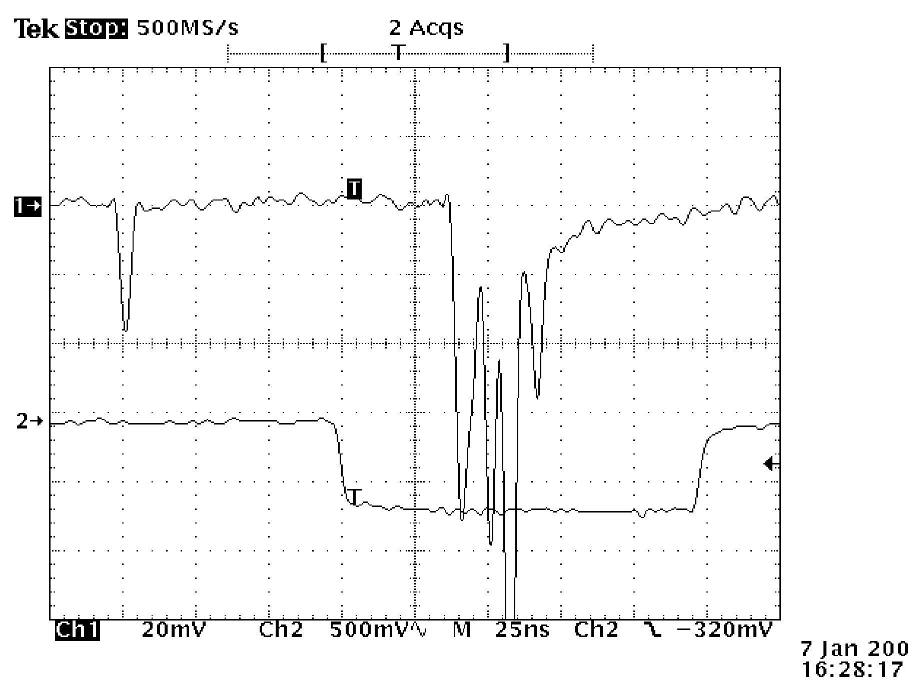
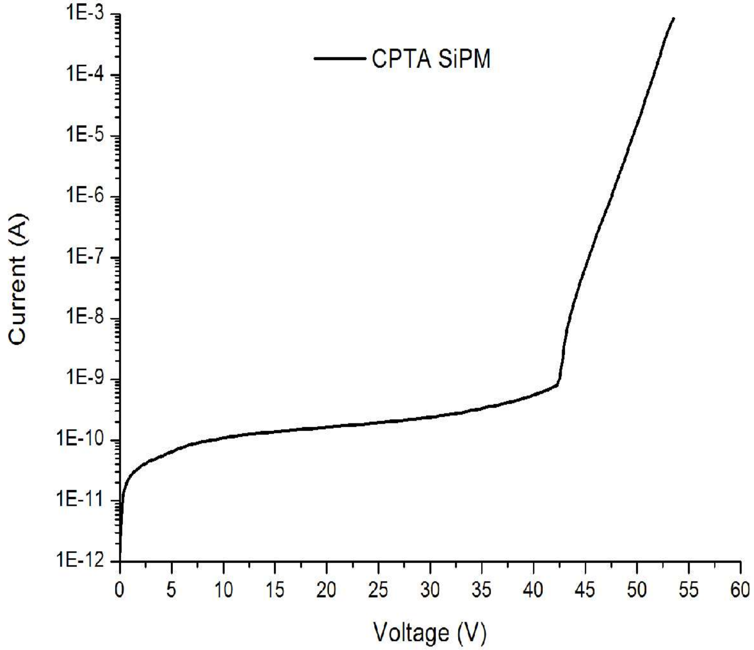




Dark current & Rate

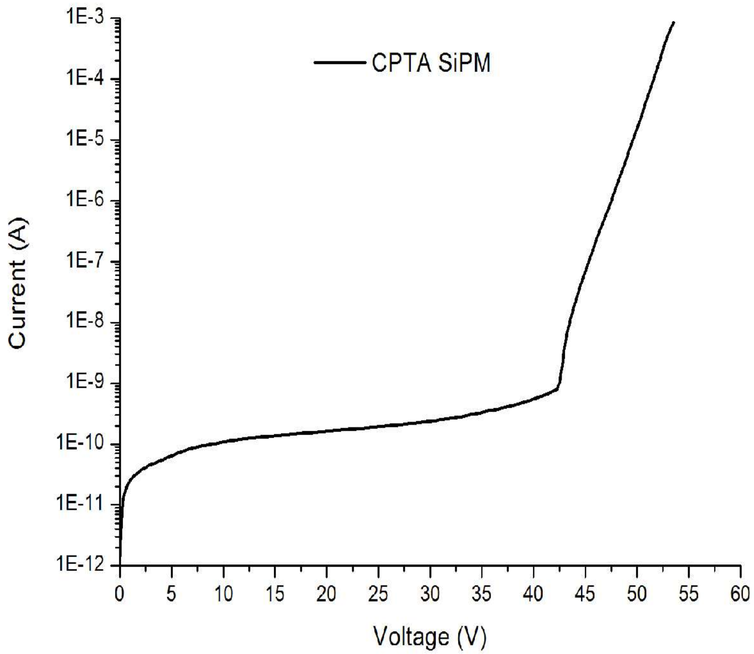


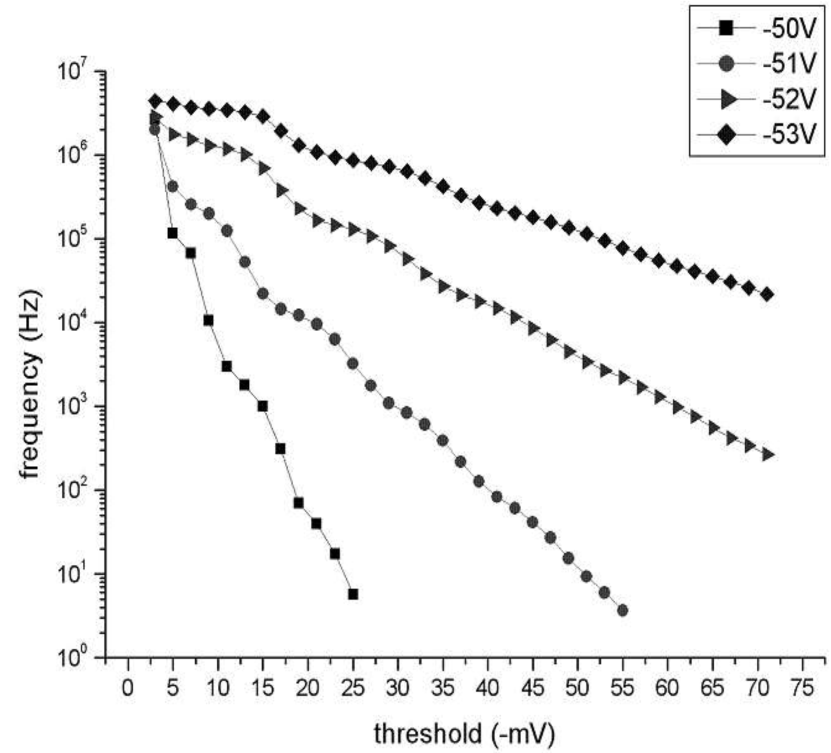
Dark counts due to...

Thermally generation of charge carriers, which depends upon

- Density of defects
- Volume

Dark current & Rate





The number of false photon counts per second registered by a SiPM in the absence of light

Dark counts due to...

Thermally generation of charge carriers, which depends upon

- Density of defects
- Volume

Optical cross-talk

- Hot carrier luminescence 10^6 carriers → 30 emitted photons in the visible range

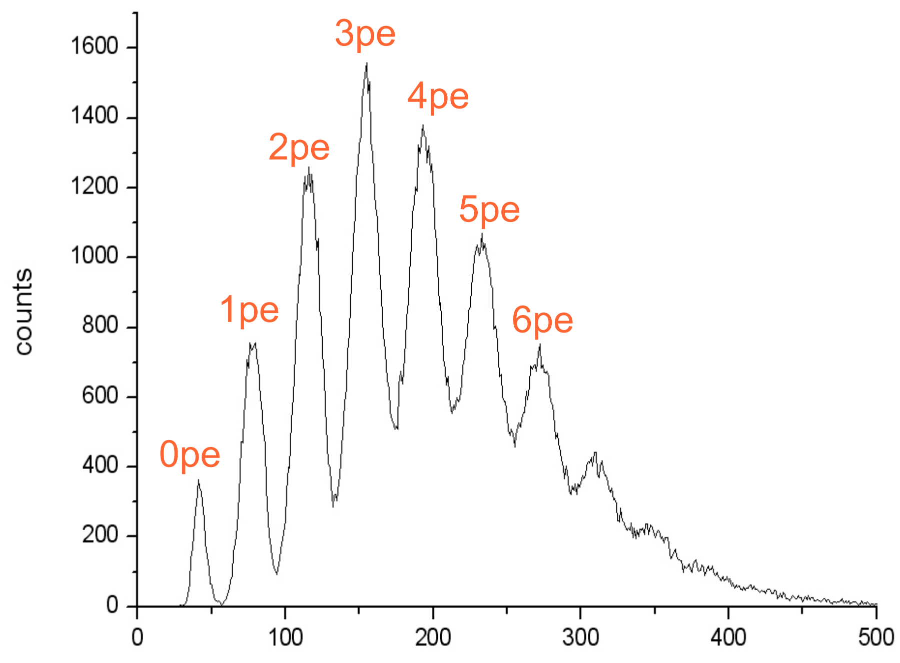
Ref: A. Lacaita, IEEE TED, vol. 40, 1993

Single Photoelectron Resolution & Gain

Gain is dependent on the capacitance of the microcells...larger cells, more gain

$$Q = C(V_{bias} - V_{breakdown})$$

Using a low light level, pulsed LED



$$Gain = \frac{Q_{one\,\mathrm{pixel}}}{\mathrm{e}}$$

Single Photoelectron Resolution & Gain

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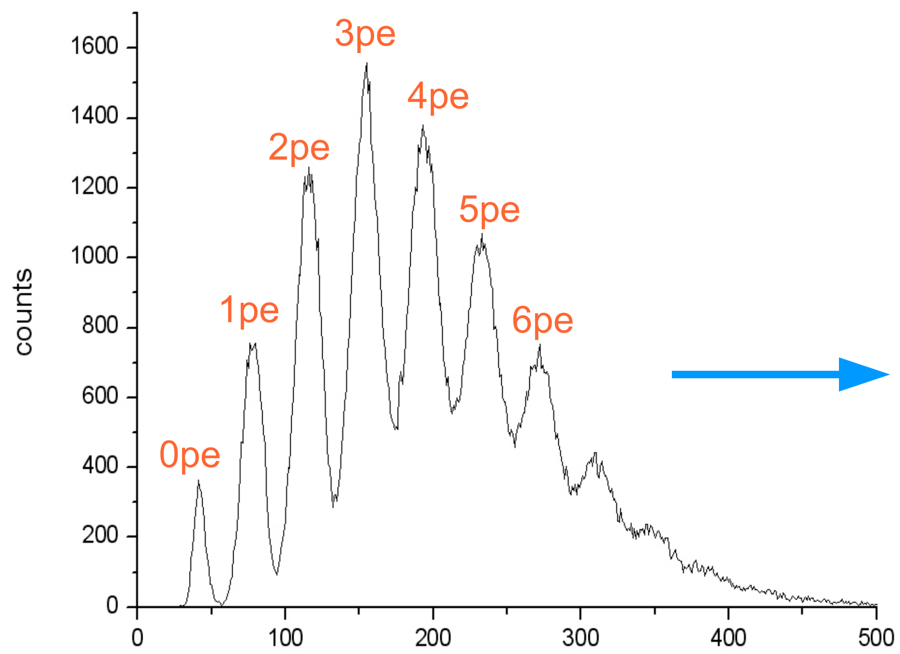
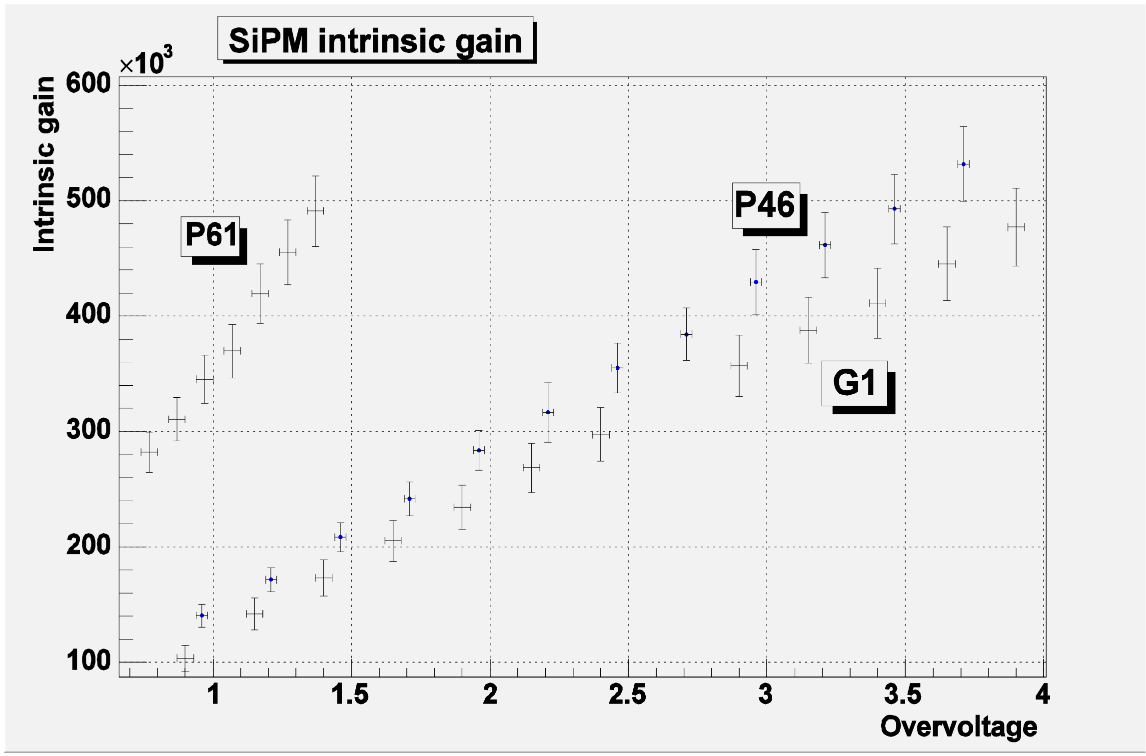
We tested with low light level, pulsed LED



$$Gain = \frac{Q_{onepixel}}{e}$$

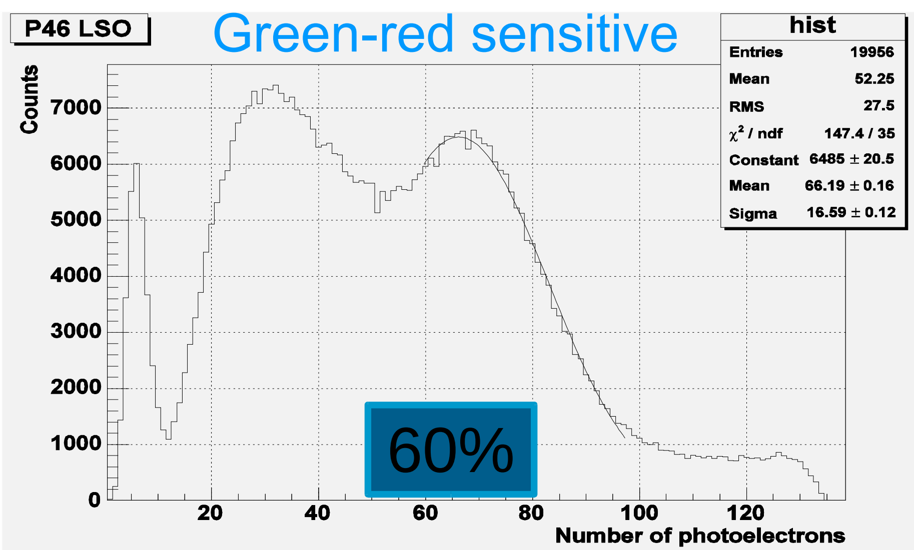
SiPM gain plotted as a function of bias

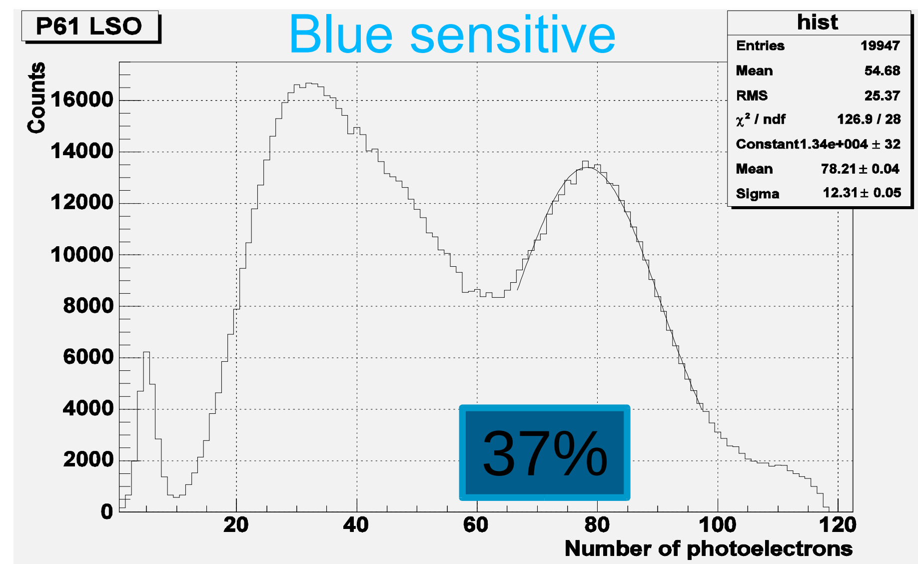


Gain is comparable to PMT

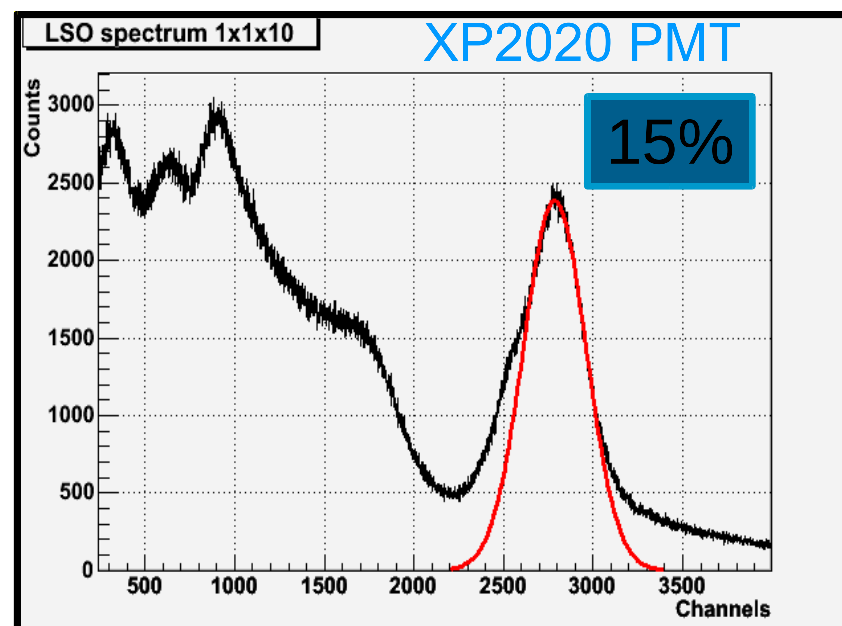
All measurements were made at *ROOM TEMPERATURE*

Scintillation readout





- SiPM coupled with a LSO crystal 1x1x10 mm^3.
- Illuminated with ^{22}Na.
- The maximum detected quantum efficiency is achieved with a high over bias – this maximises the probability of avalanche
- Even with the blue sensitive device, and high over bias, the detected light yield was found to be low – ~90 photoelectrons for a 511keV deposit.

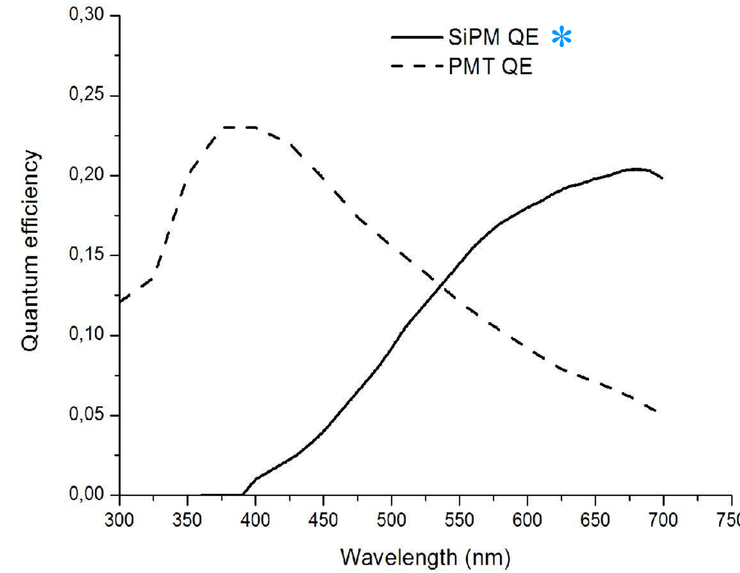


SiPM detection efficiency

Probability that a photon emitted by a light source gives an output pulse after impinging upon the detector:

$$\eta = \frac{nr.of\ output\ pulses\ recorded}{nr.of\ photons\ emited\ by\ light\ source}$$

$$\eta = \varepsilon_{geom} \times QE_{tot} = \varepsilon_{geom} \times QE \times \varepsilon_{avalanche}$$

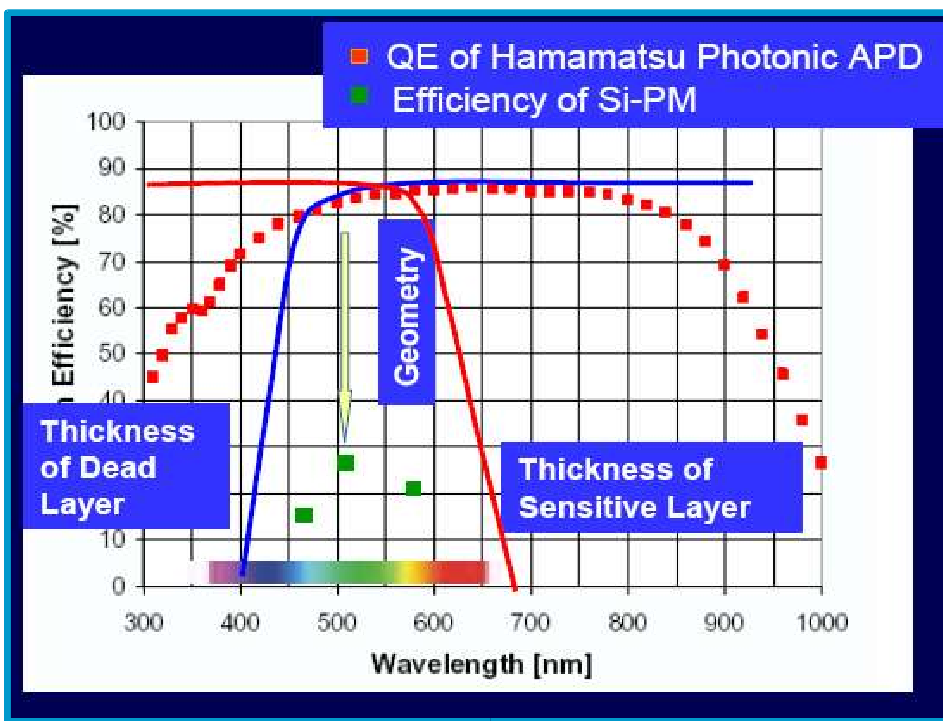


** Taken from Photonique datasheet for SSPM-040501GV-TO18*

QE_{tot} - total detected quantum efficiency :

- quantum efficiency QE
- avalanche efficiency $\varepsilon_{avalanche}$

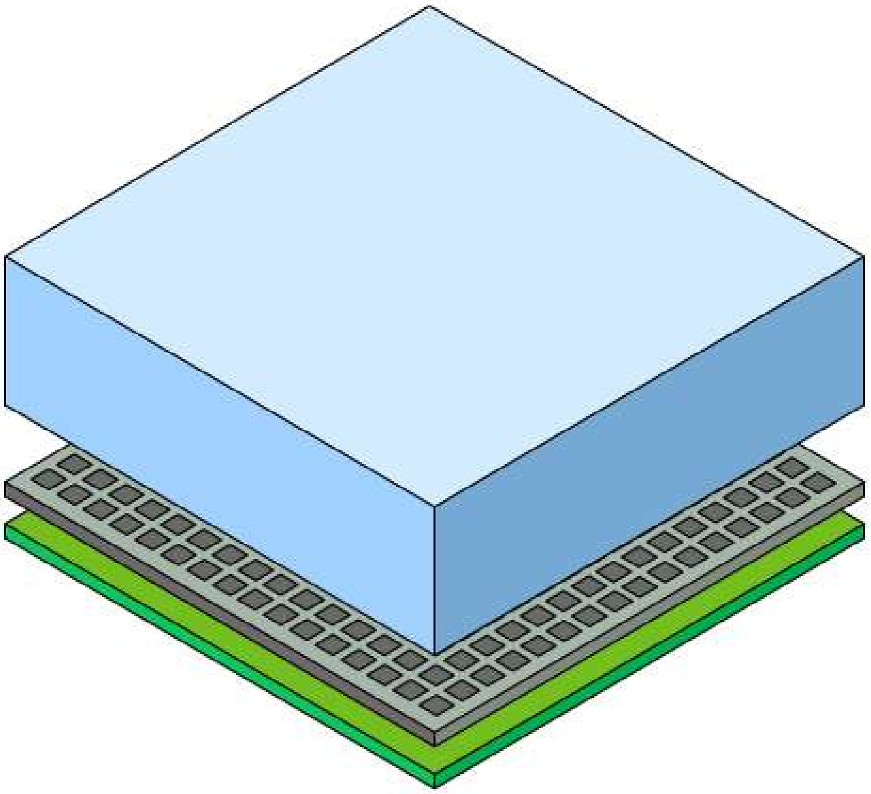
– ε_{geom} - geometrical efficiency : $\frac{\Sigma A_{pixel}}{A_{total}}$



Key element of SiPM efficiency:
geometrical factor = 0.3

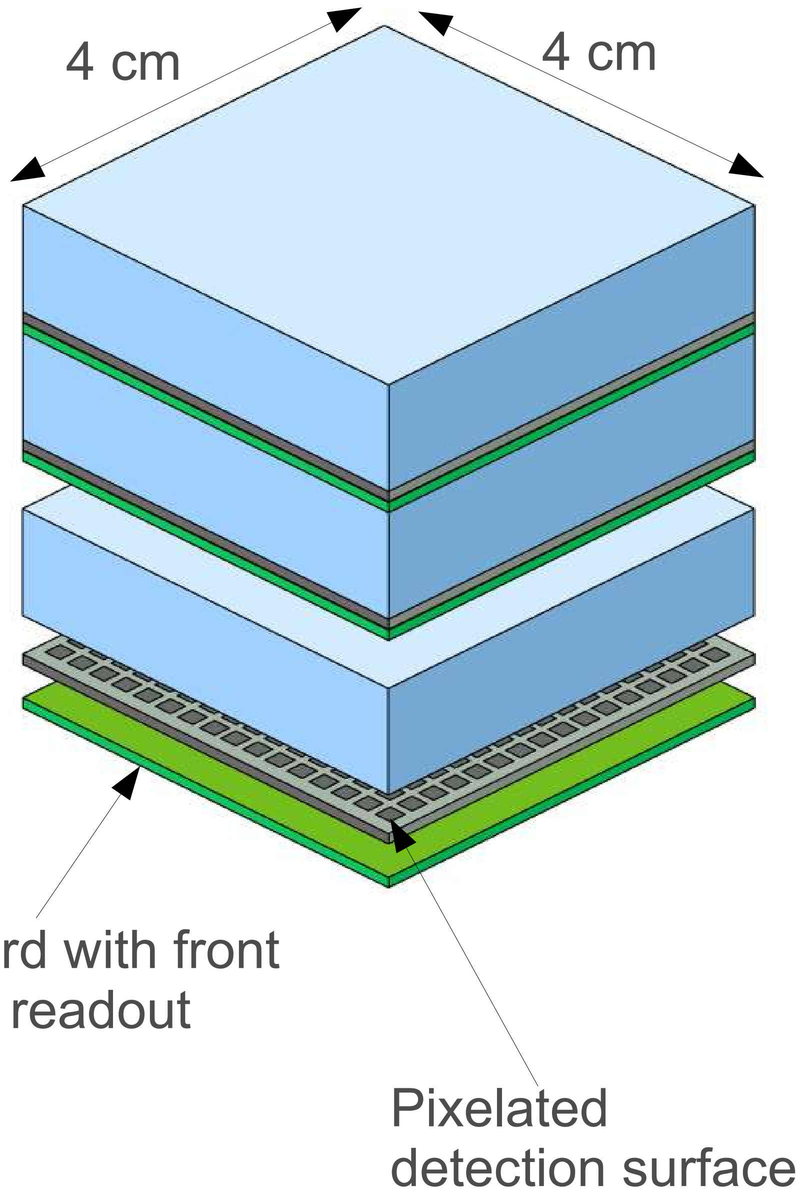
A Compact High Resolution Gamma Camera module

- Based upon principle of Anger camera.
- A continuous thin crystal spreads light over a number of detection elements.
- Center of mass of the signals give the x,y coords.
- Readout chips on board, with readout to side – compact with minimal dead space
- The data can be binned as desired, and is not fixed by the system design



Potentially very high resolution, but limited stopping efficiency

A Compact Gamma Camera Module for Small Animal PET

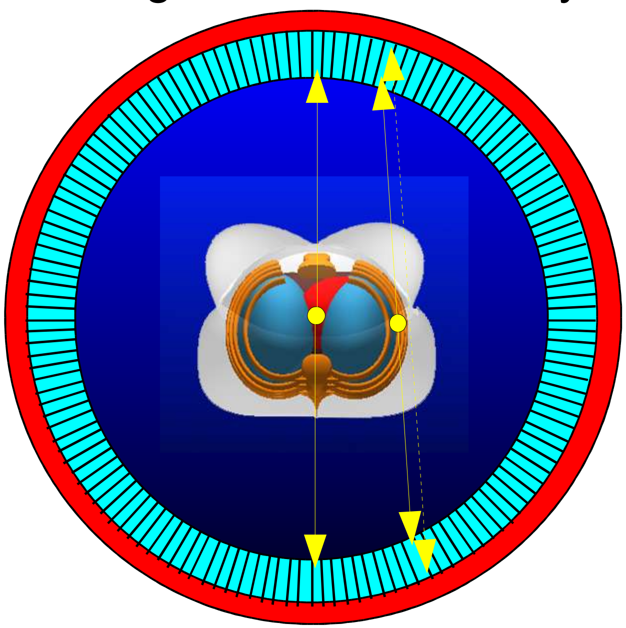


3 Layers ...

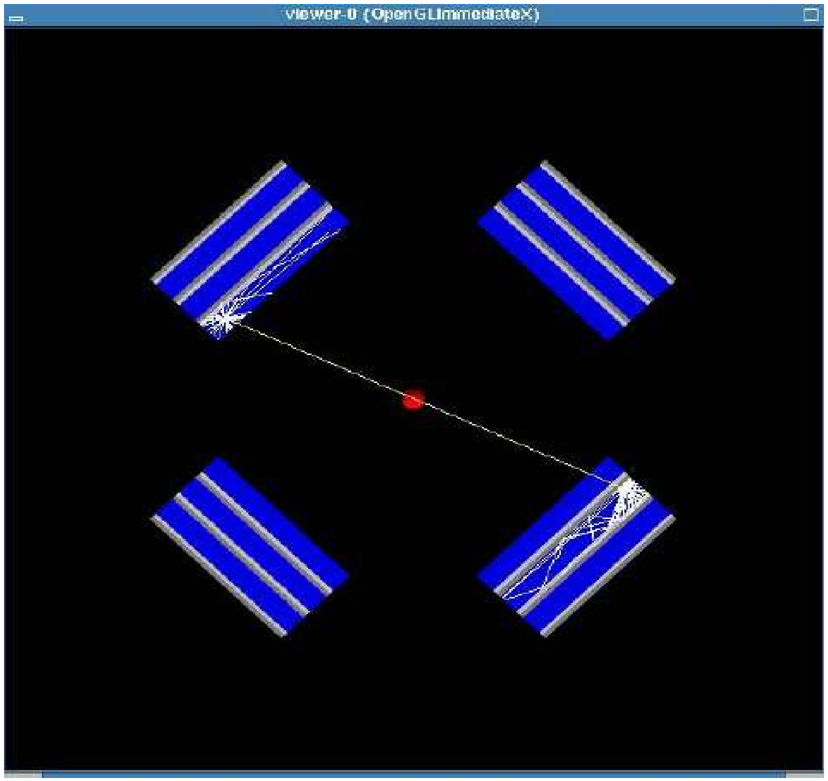
For position information we use just the *first* interaction

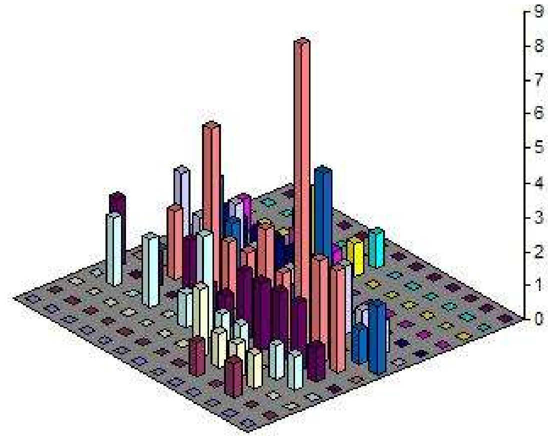
- ✔ good sensitivity
- ✔ DOI information
- ✗ but more readout

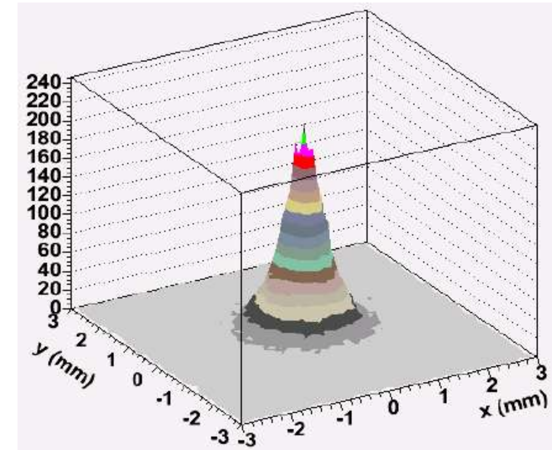
Parallax in PET: Long crystals give positioning errors for skew rays

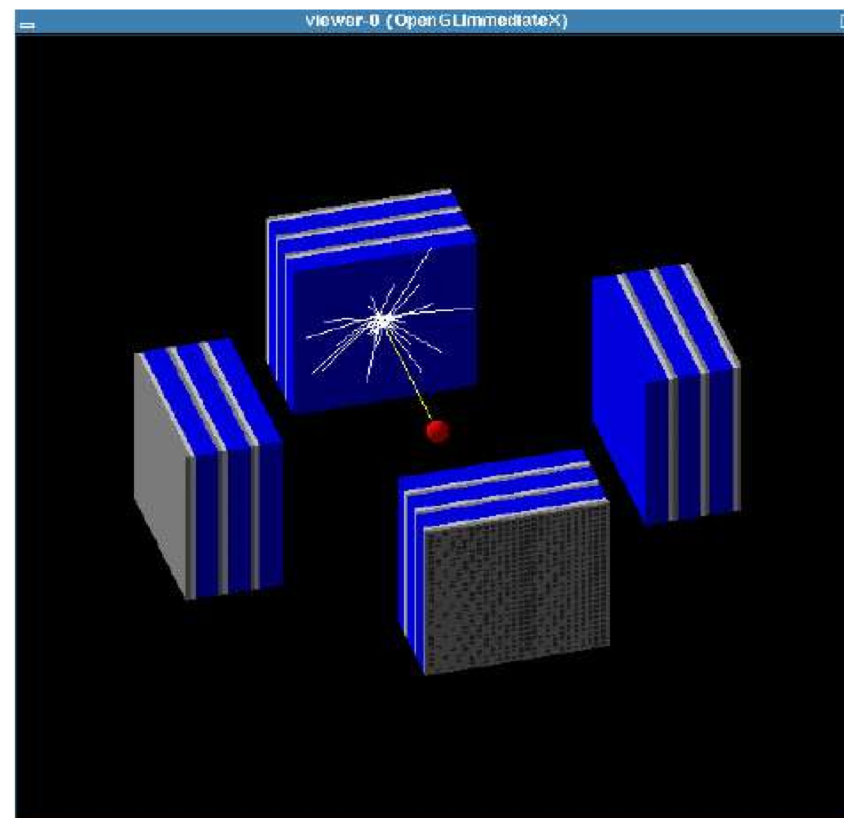


Simulation Set up



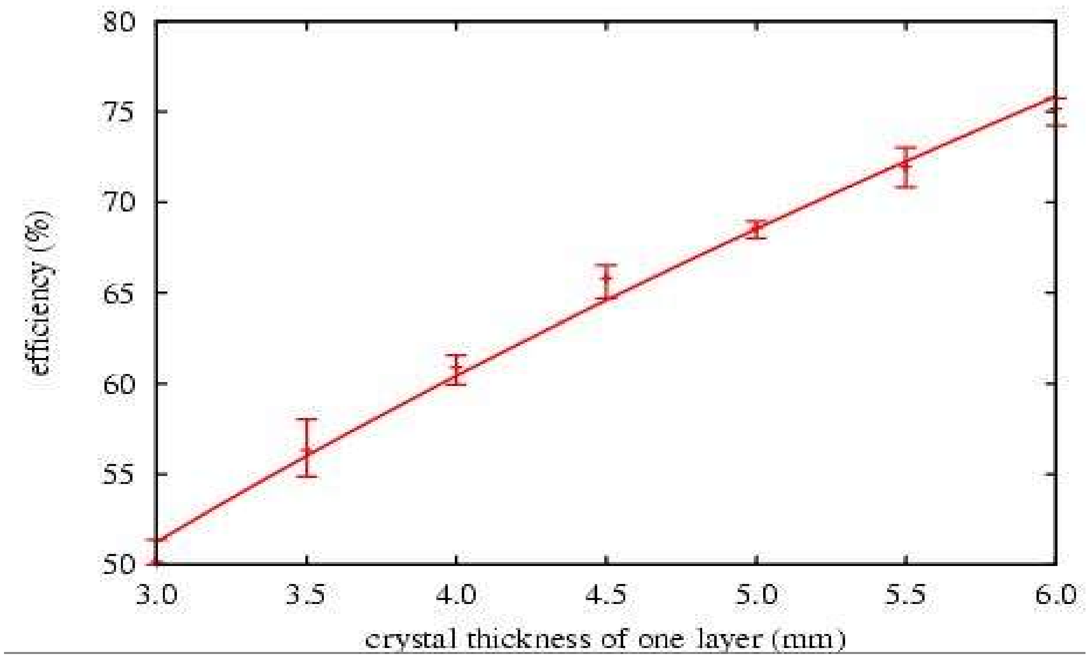






- Simulations made with GEANT 4
    - parallel beam of 511keV gammas
    - 50keV threshold
    - scintillation photons generated
    - QE used averages 2.5% over LSO emissions
    - SiPMs: $1x1mm^2$ on a 1.5mm pitch

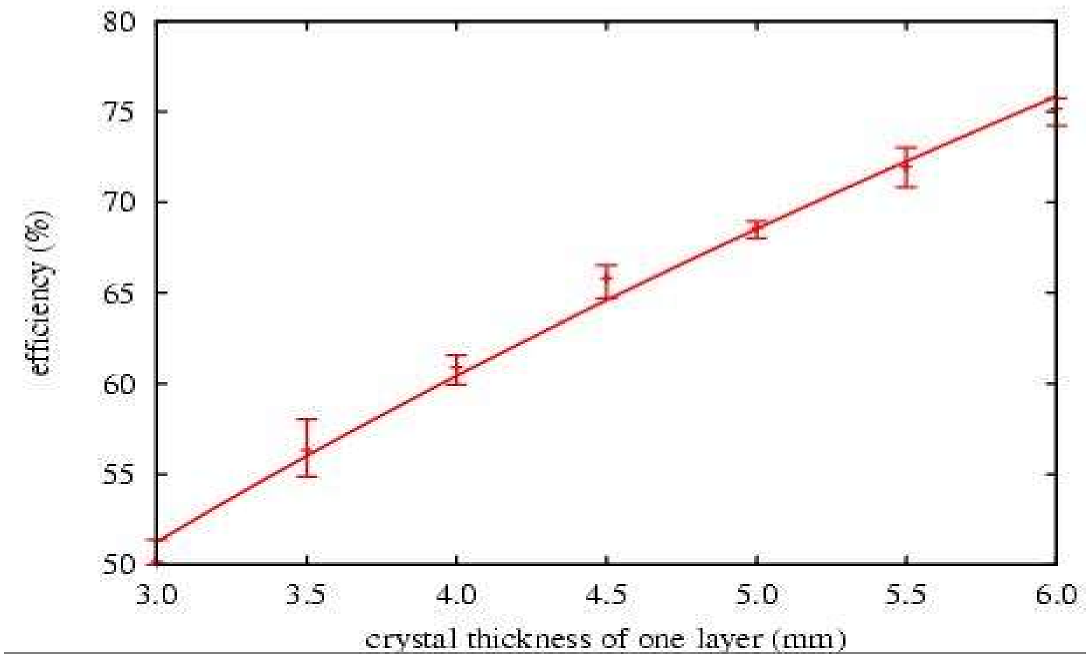
Efficiency, Backscatter, Count rate



Efficiency -

- Percentage of interactions depositing > 50keV in a single layer
- Simply a function of thickness

Efficiency, Backscatter, Count rate



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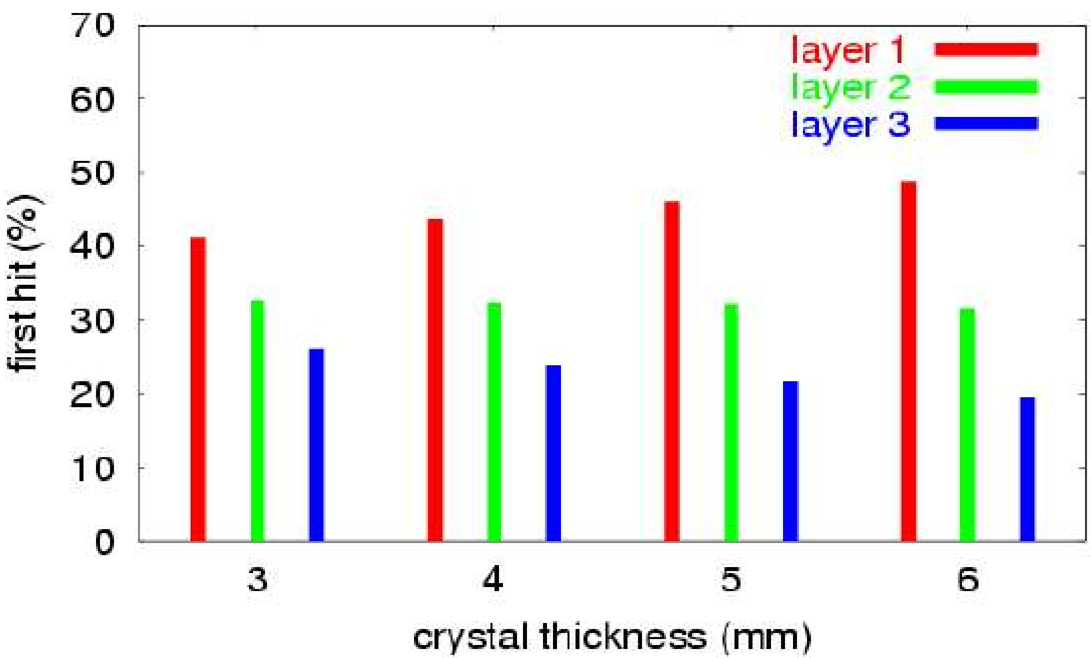
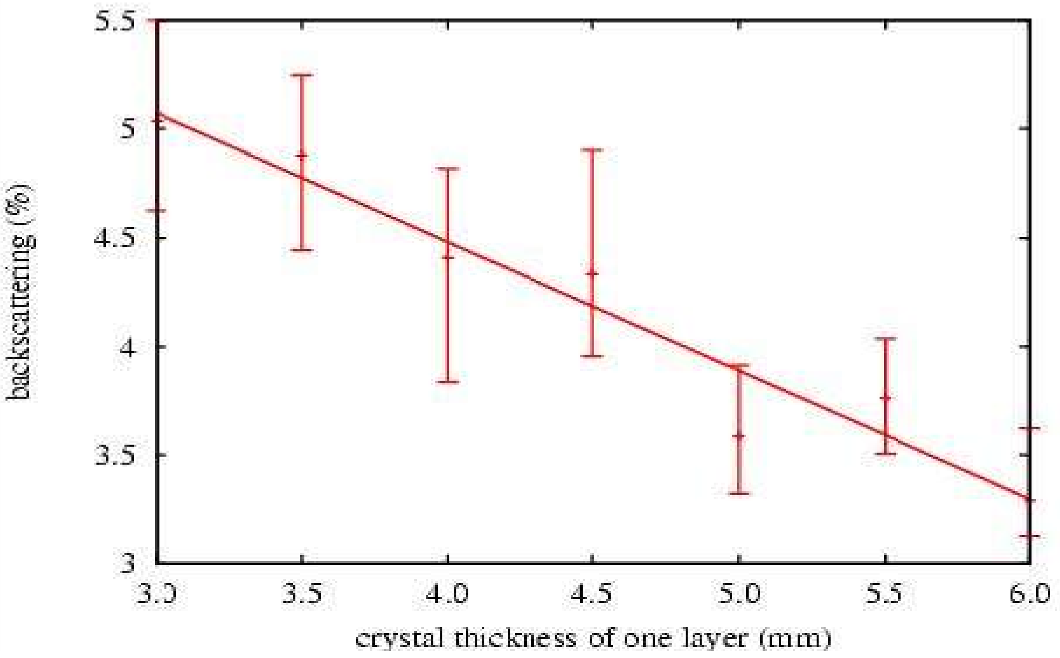


Counts -

- Number of 'first' interactions in each layer
- More counts in the front layer
- The disparity in count rate increases with thickness
- For 6mm layers, the count rate is 2.5 times that of the back

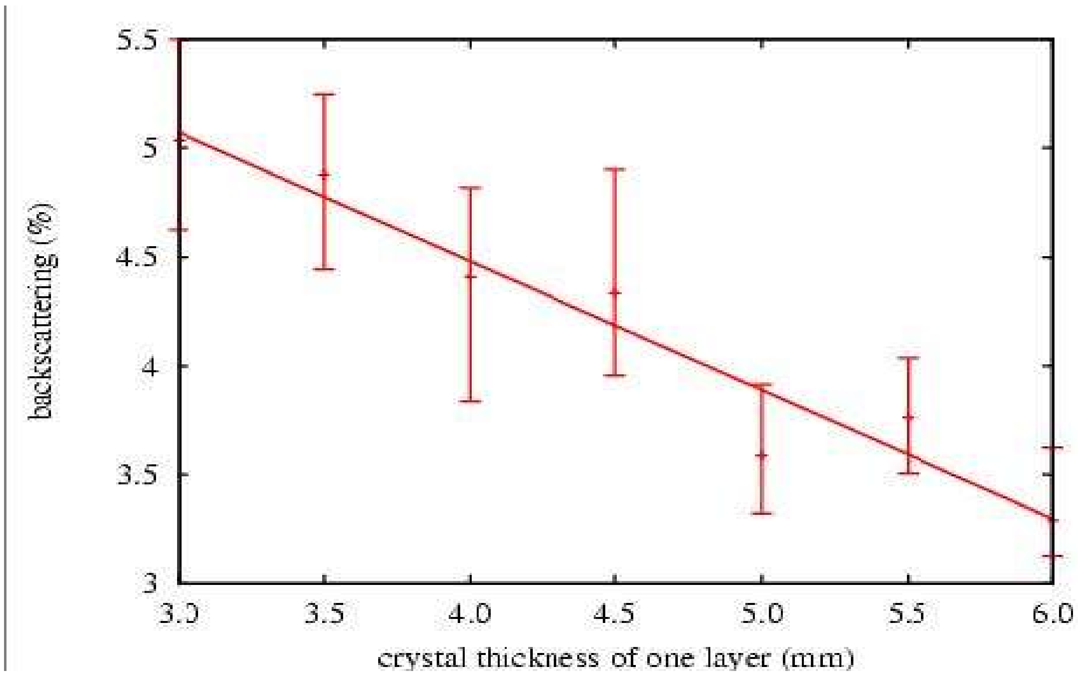
Back scatter & Spatial Resolution

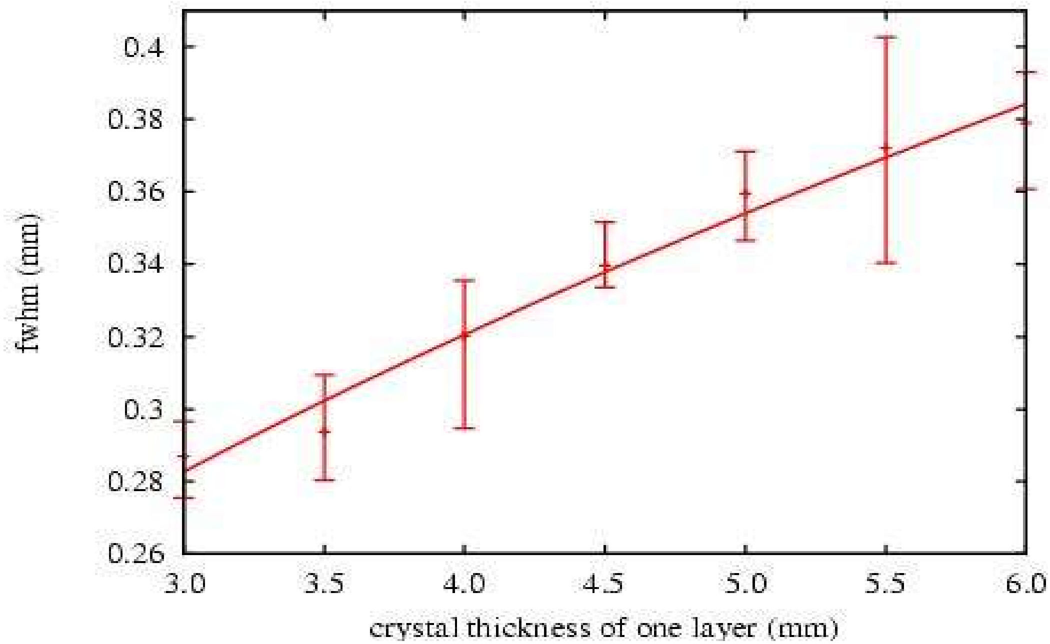


Back scatter -

- Percentage of interactions that deposit energy and scatter into a layer nearer to the FOV and deposit energy
- This results in incorrect DOI selection
- The probability increases with decreasing crystal thickness but still only 5%.

Back scatter & Positioning Accuracy





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Positioning accuracy -

- Event positioning accuracy – FWHM of event distribution
- Well below sub millimetre.
- The dependence on crystal thickness is not strong, leading us to prefer a thicker crystals for the extra sensitivity

Other Resolution Considerations

Parallax contribution

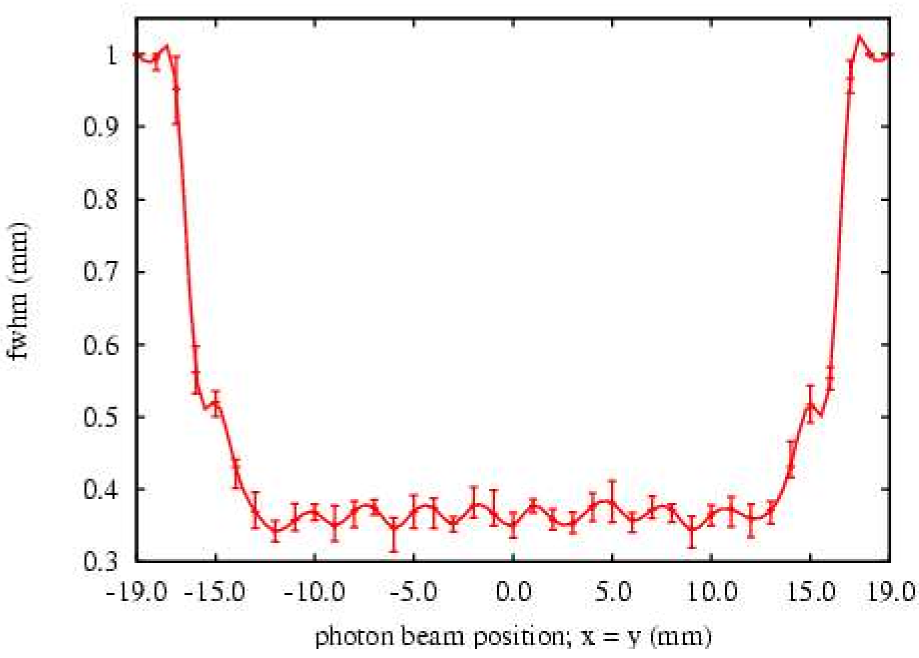
- A thick (5mm) crystal was seen to give good sensitivity and only a small decrease in the FWHM resolution.
- But implies worse DOI resolution which degrades the intrinsic resolution.
- Using a geometrical argument, with skew rays at the maximum of 22°, the contribution to the error in x or y, from the uncertainty in z is ±1mm.

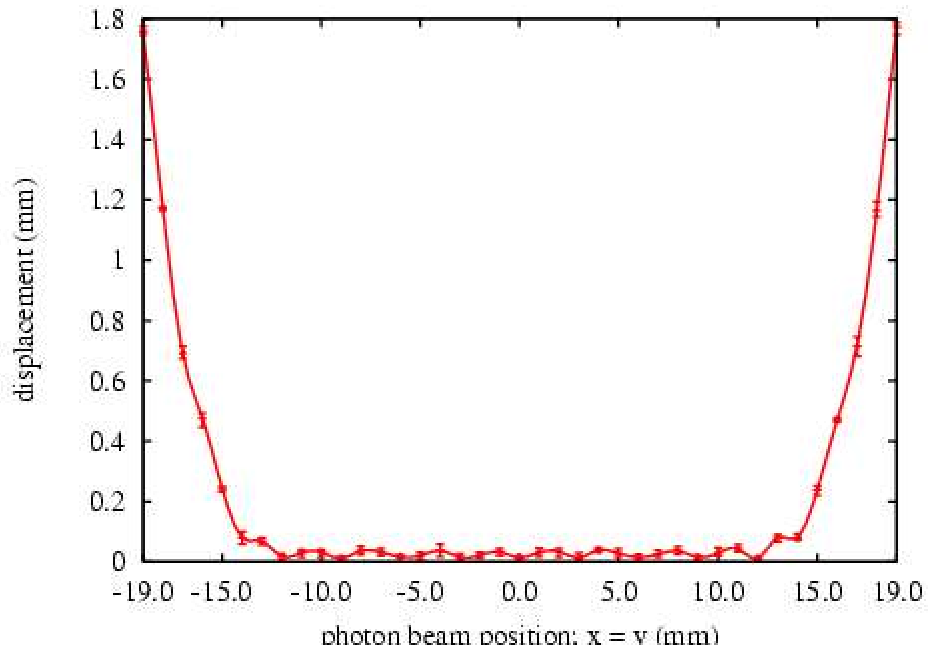
- Additional contributions to the image resolution in PET are
    - The positron range (0.5mm)
    - Acolinearity (< 1mm)
    - DOI (1mm?)
    - Intrinsic camera resolution (< 0.5mm)

Other Resolution Considerations

Edge distortions

- As in all gamma cameras that reply on Anger principle, the resolution and positioning accuracy is degraded towards the edges, as shown below.
- Possible position information recovery;
    - Skewness
    - Learning methods





Summary I

*The **SiPM** offers a very promising opportunity for creating compact, modular and low cost, application specific gamma cameras.*

- Possible replacement for PMT?
- Benefits of compactness, ruggedness, low bias, simple electronics, unaffected by magnetic fields
- Need QE optimised for blue-green, matrices development and compact electronics

Current SiPM development

Many rapid developments, ultimately heading towards matrices

- **Photonique (CH)**
- **Hamamatsu**
- **SensL (IRL)**
- **ITC-IRST (IT) (MEMs / SiPM funded by INFN)**

Summary II

*The SiPM offers a very promising opportunity for creating compact, modular and low cost, **application specific gamma cameras.***

- Many configurations can be realised with a SiPM detector.
- Use of a continuous crystal reduces costs and provides continous sampling giving better quantification and flexible data binning.
- A simple camera design can yield both high spatial resolution and high sensitivity.
- Resolution recovery solutions for the edges of continuous cameras can be implemented.

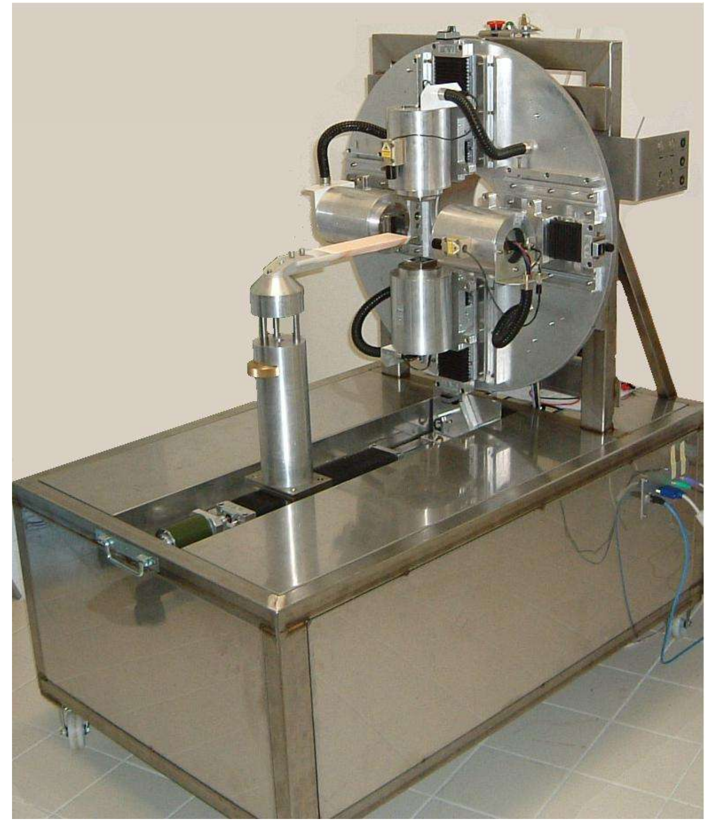
- *How could such a system be adapted for Prostate imaging?*

End

YAP- (S)PET - An integrated PET / SPECT small animal scanner

Scanner configuration

Configuration:	Four rotating heads
Scintillator:	$YAlO_3$:Ce (YAP:Ce)
Photodetector:	Position Sensitive PMT
Readout method:	Resistive chain (4 channels)
FoV size:	4 cm axial 4 cm Ø
Collimators: (SPECT)	Lead (parallel holes)
Animal bed:	Motorized, PC controlled
Animal positioning:	Laser pointers



YAP-(S)PET performance

PET mode:

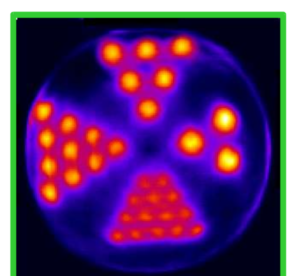
Spatial resolution (iterative algorithm)
Volume: 5.2 mm^3 (4.4 mm^3) @ CFOV

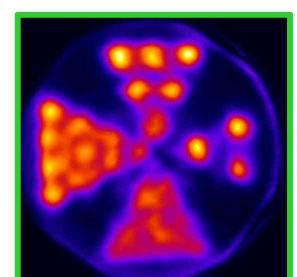
Sensitivity
19 cps/kBq @ CFOV (50-850 keV)

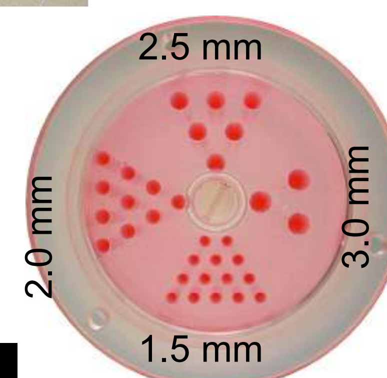
SPECT mode:

Spatial resolution (iterative algorithm)
Transaxial: 3.1 mm (R) × 3.9 mm (T)

Sensitivity
30 cps/MBq (constant)







Images of a mini-Derenzo phantom obtained in PET and SPECT modalities (EM reconstruction). Left: PET (18F, 50-850keV energy window), right: SPECT (99mTc, 140-250keV energy window) 11 2 mm^3 voxel space (no contrast enhancement).