The Silicon Photomultiplier for Application to High-Resolution Gamma Cameras for PET

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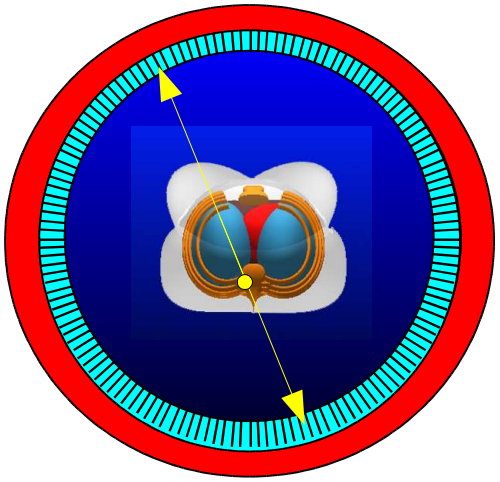
Position Sensitive Detectors 7, 13th September 2005, Liverpool

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

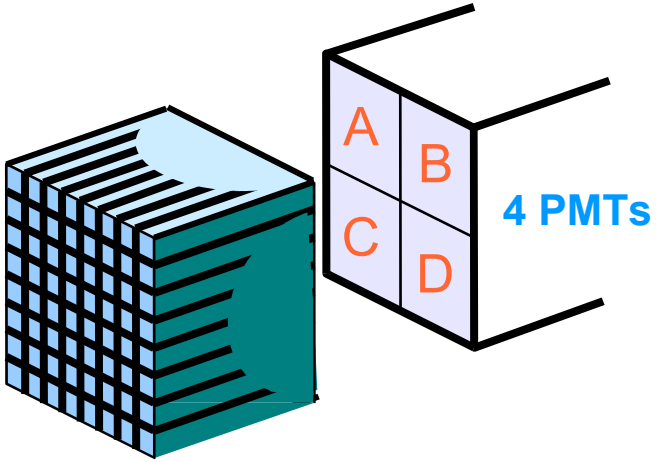


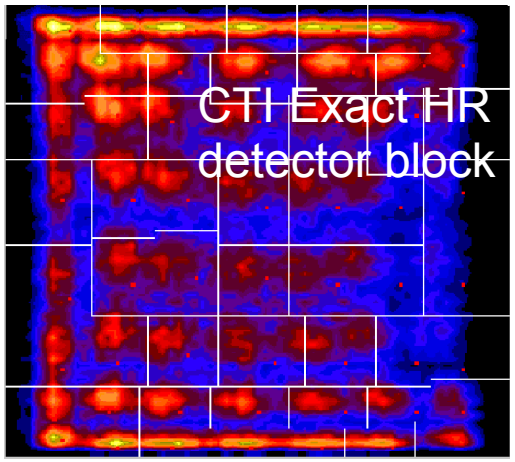
- ***PET camera requirements***

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

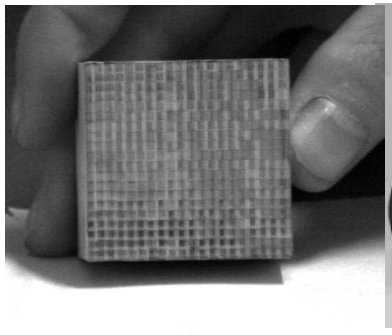
Detectors for PET

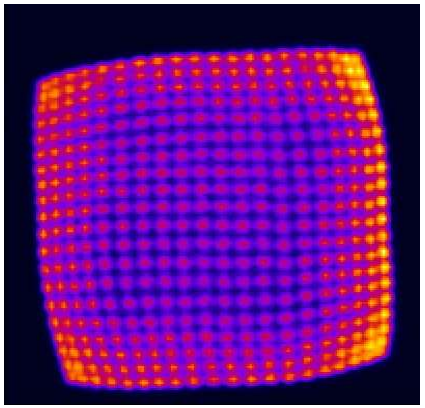
- Classic block detector (BGO then LSO) is still used but *limited in spatial resolution*



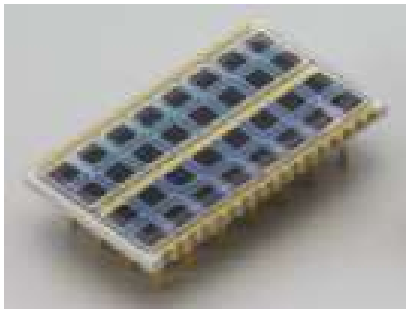


- For higher resolution, PSPMTs and pixelated scintillator arrays are used but they are of *limited size and expensive.*



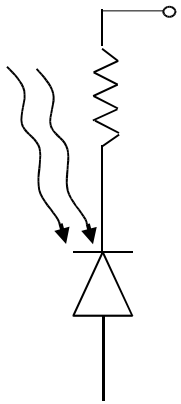


- Now people are looking to arrays of semiconductor detectors- i.e.) APD matrices.
  - *limited gain*
  - *limited size*

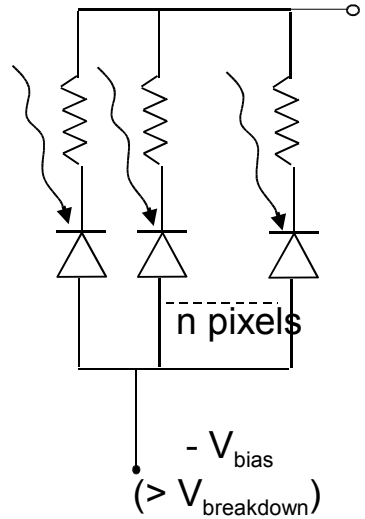


A new detector – Silicon Photomultiplier (SiPM)

GM-APD + $R_{quenching}$



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



SiPM

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

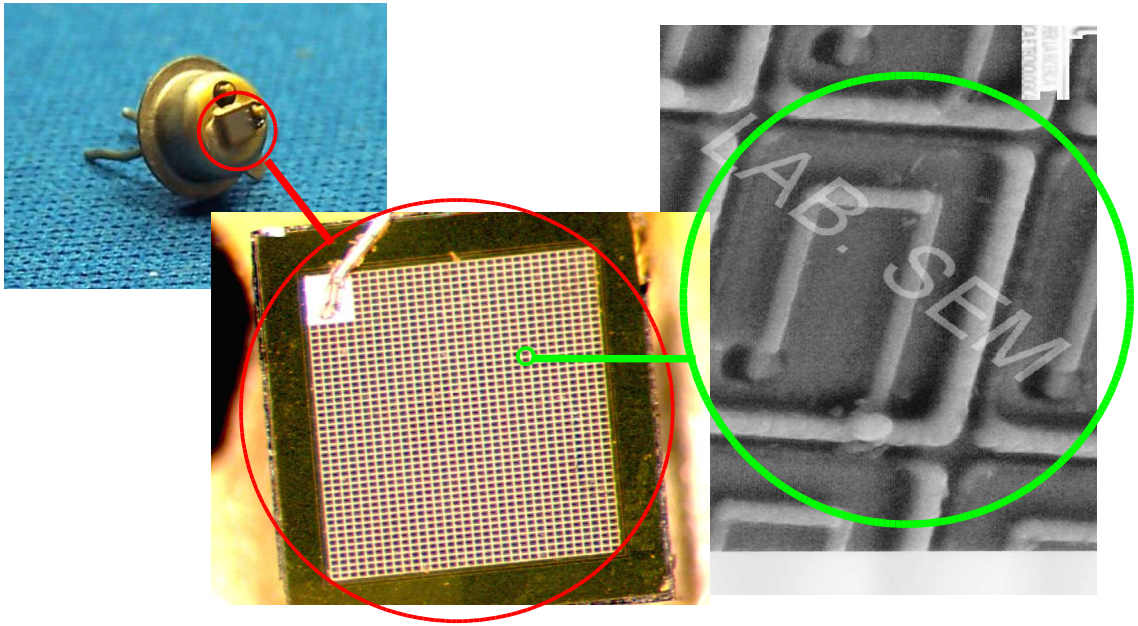
Output signal is the sum of the signals from all fired pixels *(analogue device)*

Possible replacement for vacuum PMT?

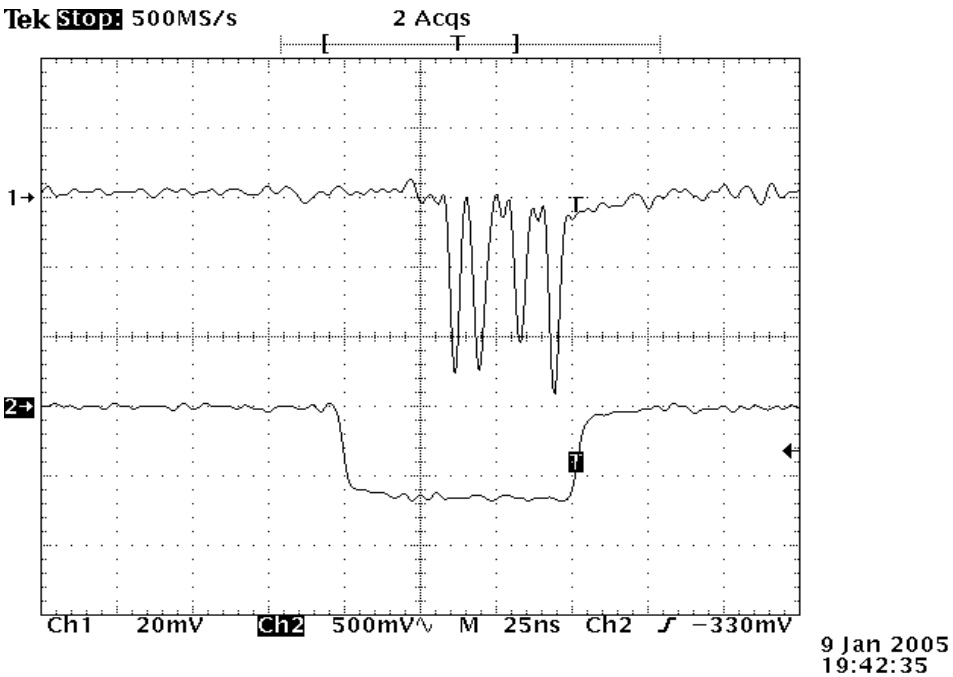
- ✔ High internal gain (10^6) at low bias voltage (~50V)
- ✔ Excellent photon counting capability
- ✔ Fast response time (~1ns)
- ✔ Noise (dark counts) limited to photoelectron level
- ✔ Insensitivity to magnetic fields
- ✔ Compact and rugged

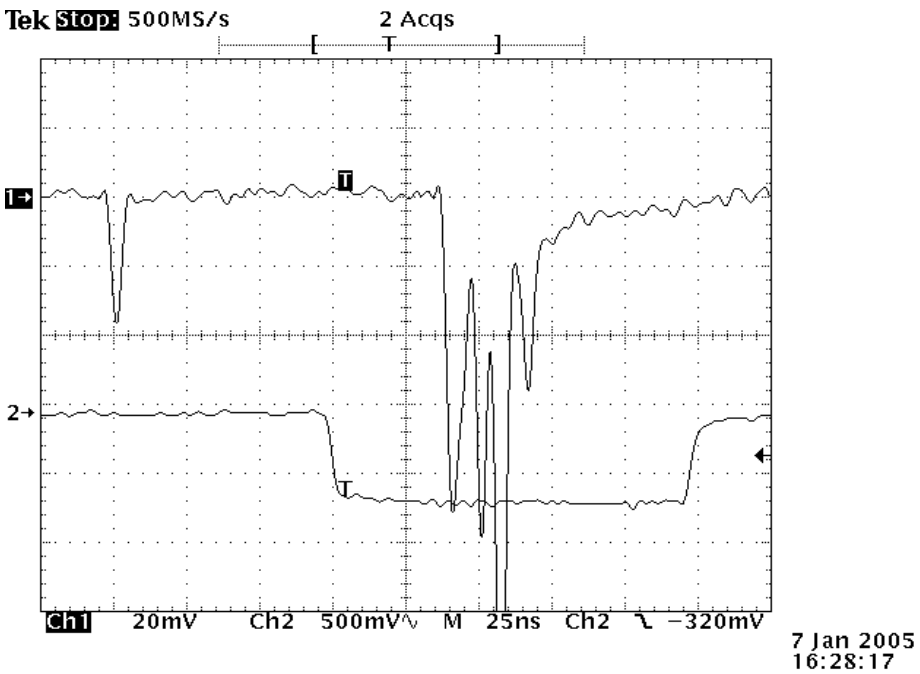
- ✗ Low detection efficiency (ave. 2.5% over LSO emission)
- ✗ Limited dynamic range (1000/mm^2)
- ✗ Not yet available in matrices

SiPM testing

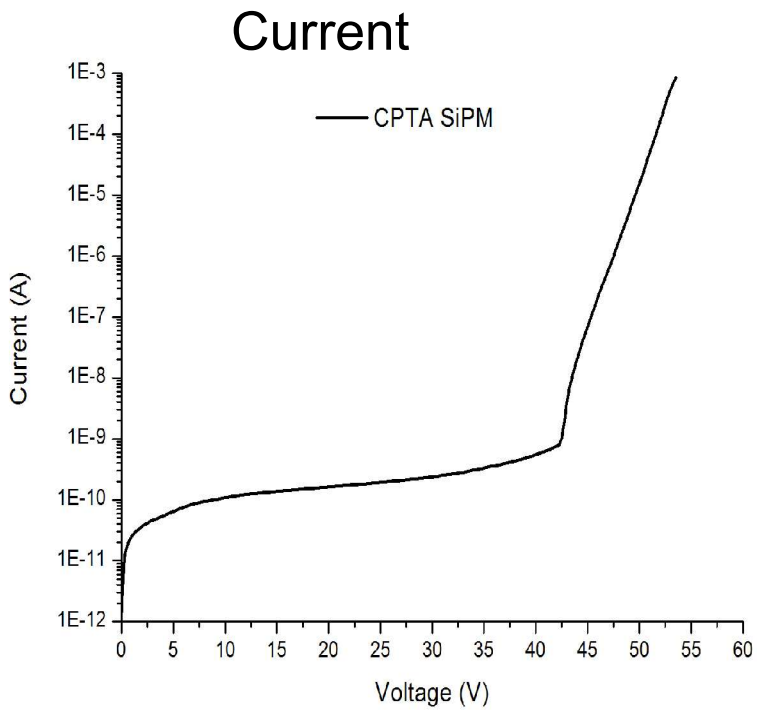


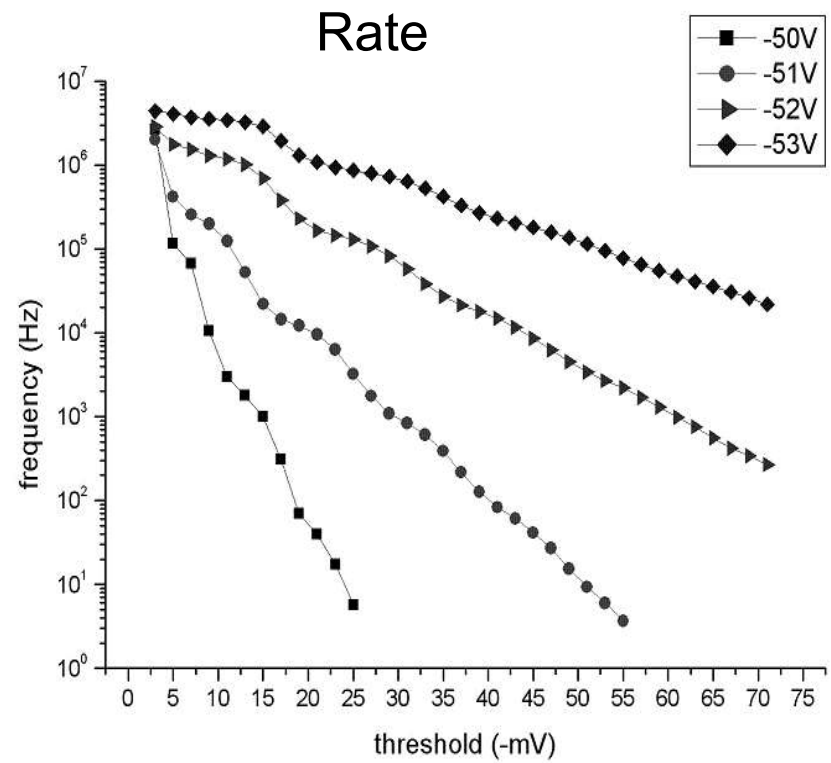
- Samples from CPTA, Russia
- ~ 50V
- Fast preamp + QDC





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

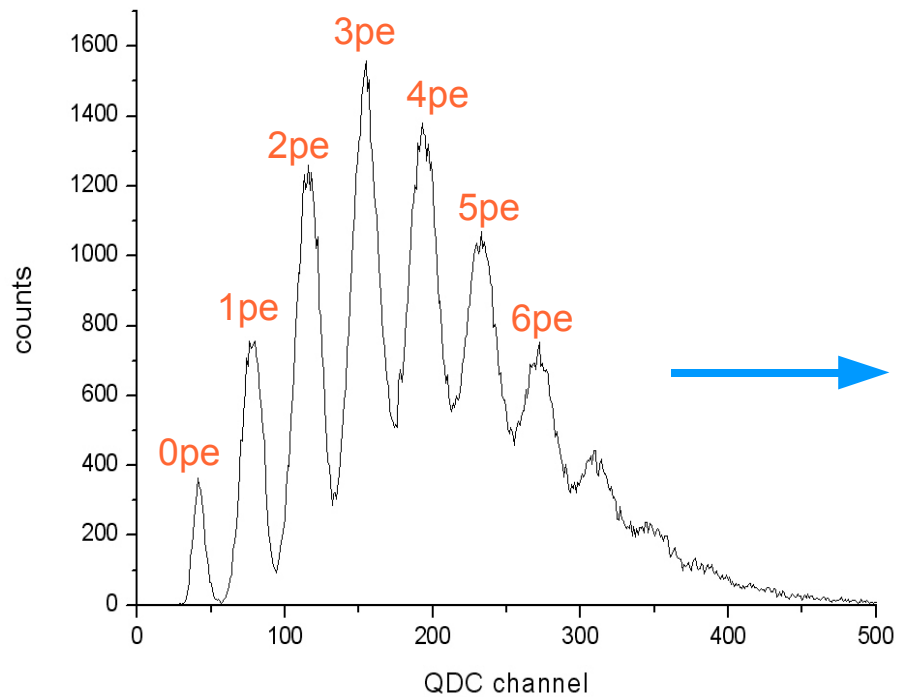
- Depends on the density of defects
- Volume

Multiple photoelectron structure

- Evidence of optical cross-talk

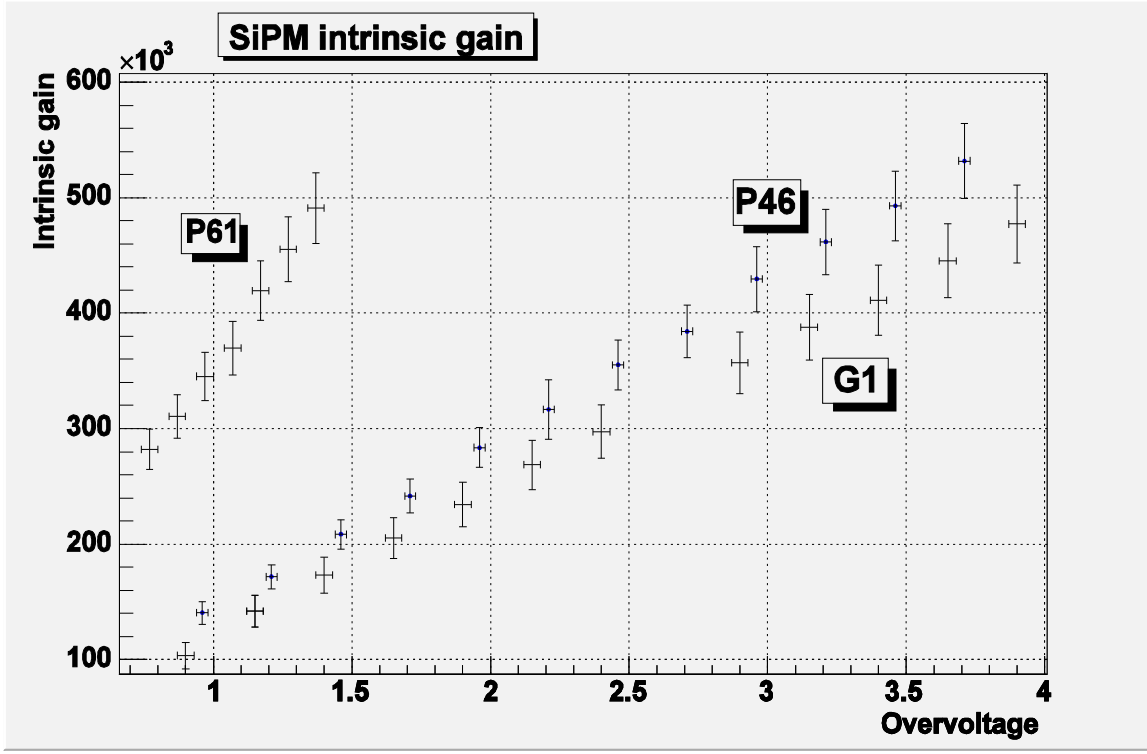
Single Photoelectron Resolution & Gain

We tested with low light level, pulsed LED



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

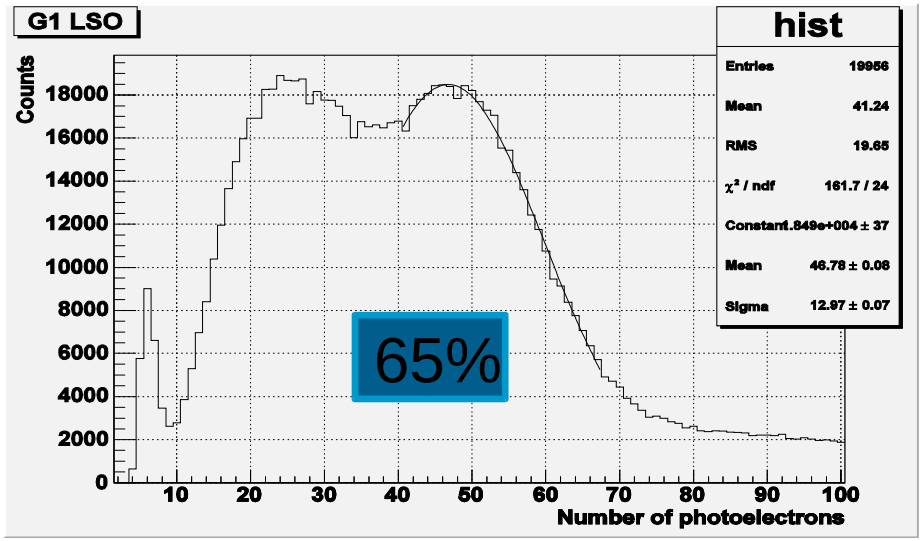
SiPM gain plotted as a function of bias

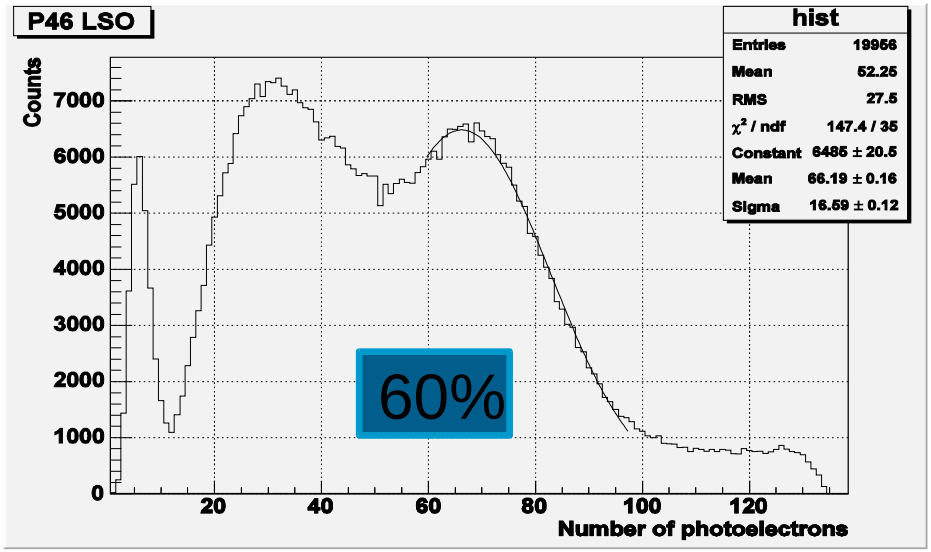


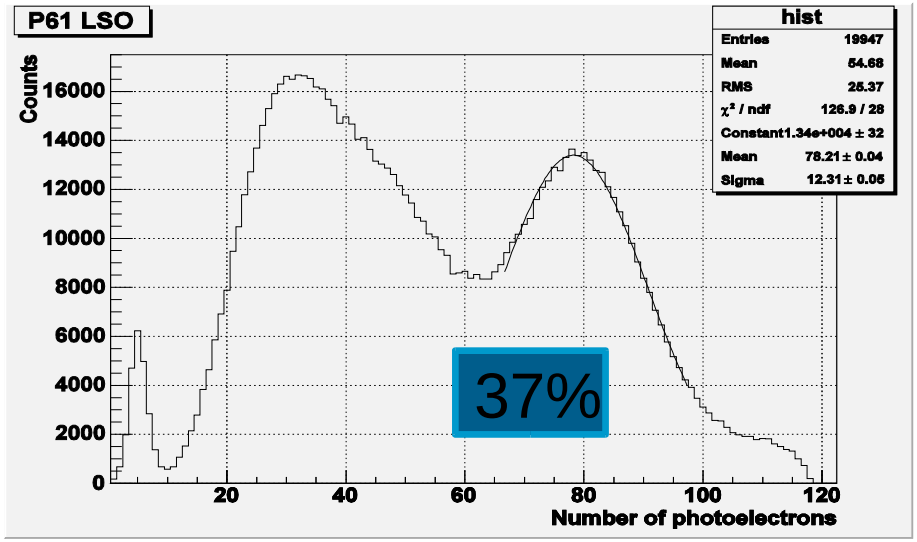
Gain is comparable to PMT

All measurements were made at *ROOM TEMPERATURE*

Scintillation readout & Quantum Efficiency



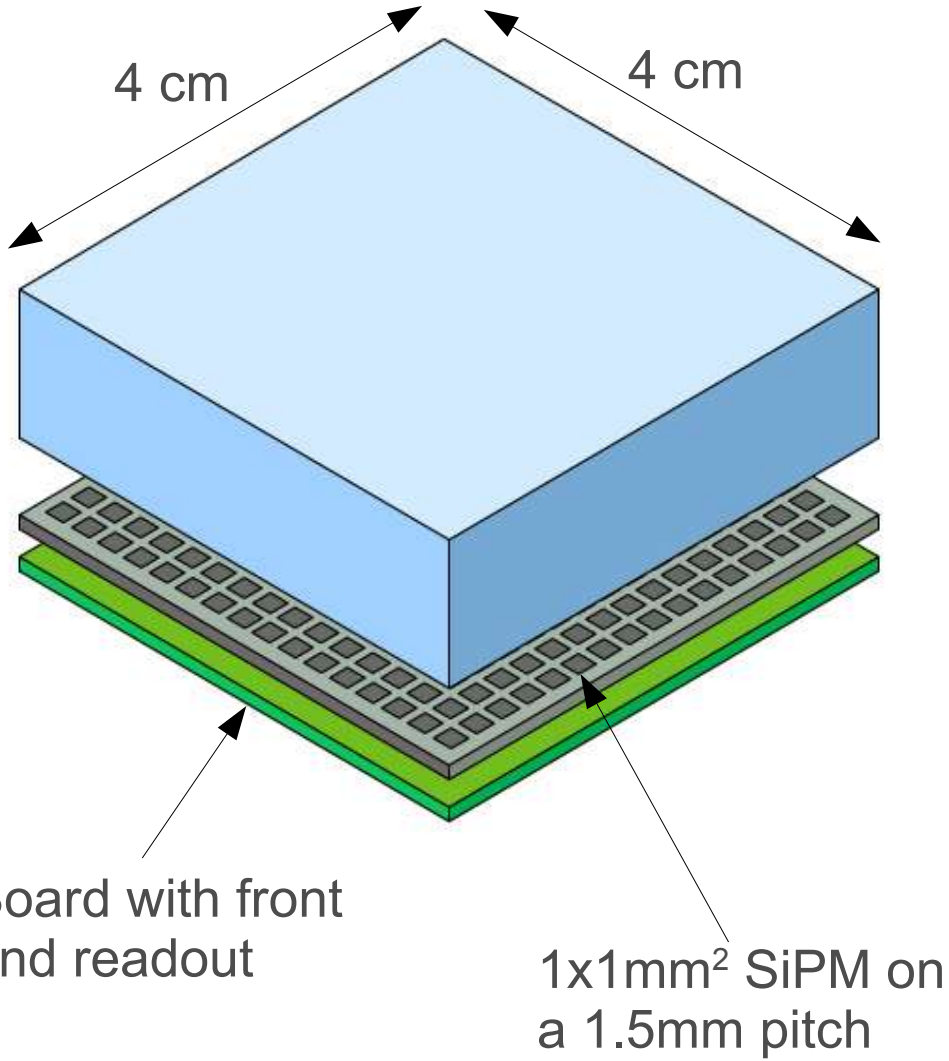




- SiPM coupled with a LSO crystal 1x1x10 mm.
- Illuminated with ^{22}Na.
- Low quantum efficiency!!!

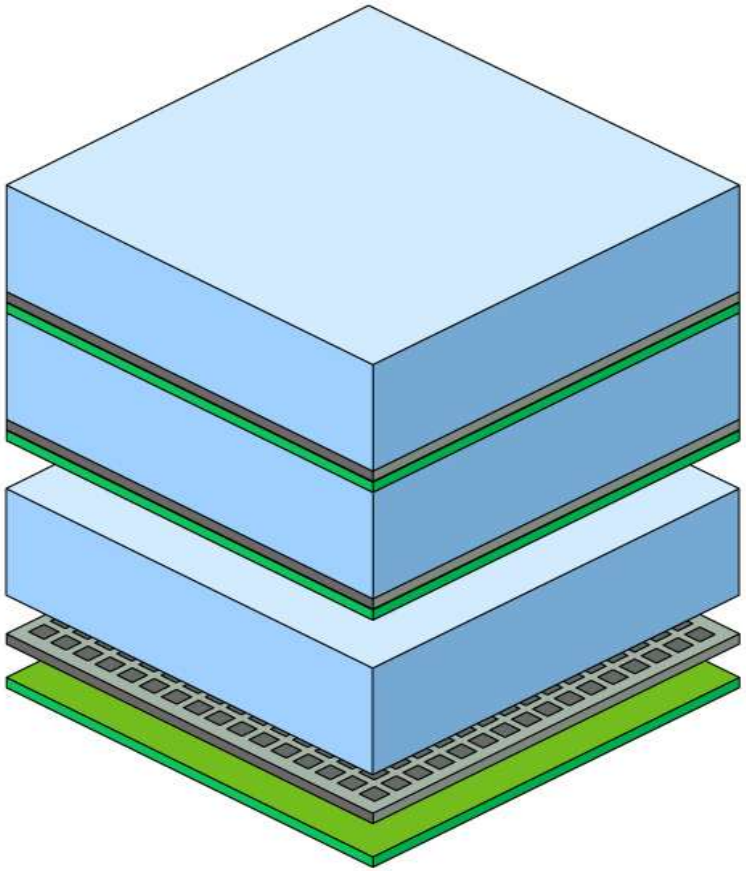
A Gamma Camera based upon SiPM

- 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



Potentially very high resolution, but limited stopping efficiency

A Gamma Camera based upon SiPM

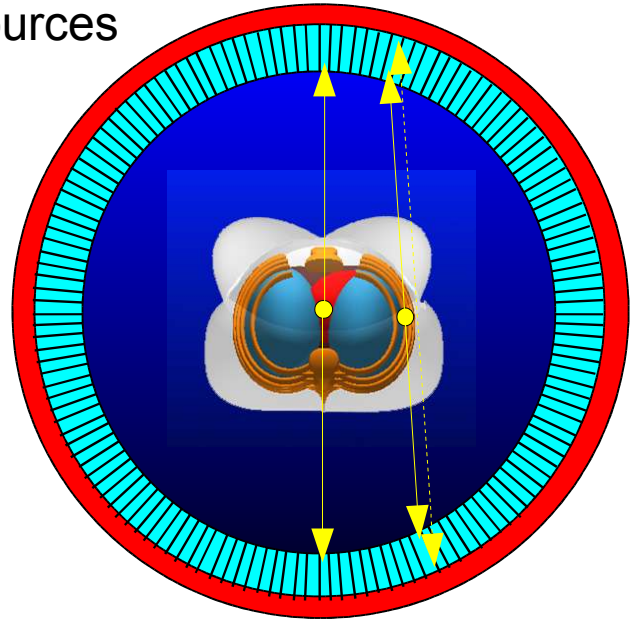


3 Layers ...

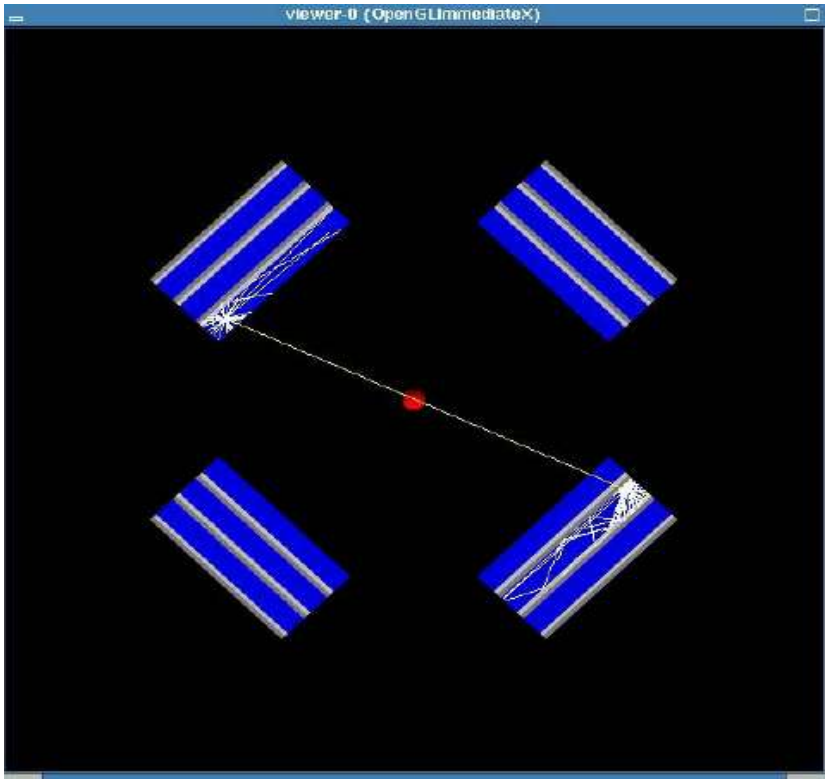
For position information we use just the *first* interaction

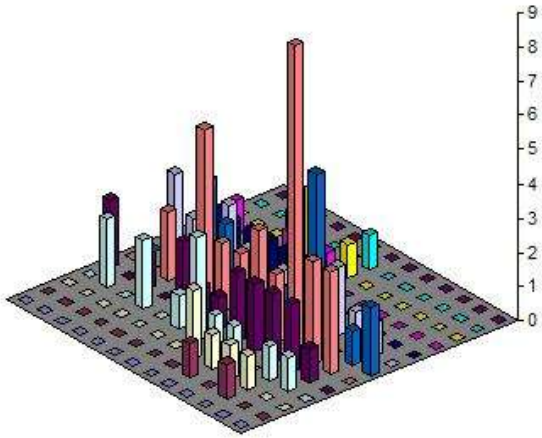
- ✔ good sensitivity
- ✔ DOI information
- ✗ but more readout

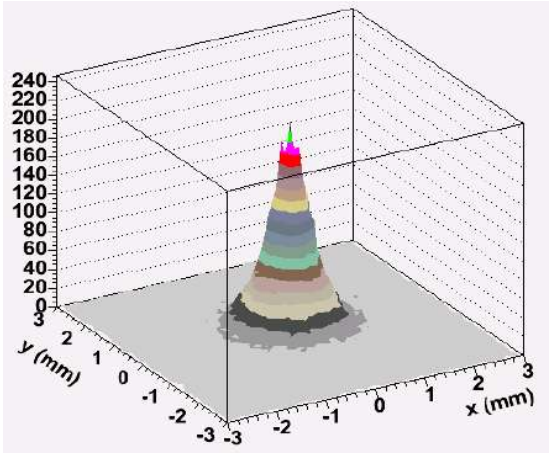
Parallax in PET: Long crystals give positioning errors for off-centre sources

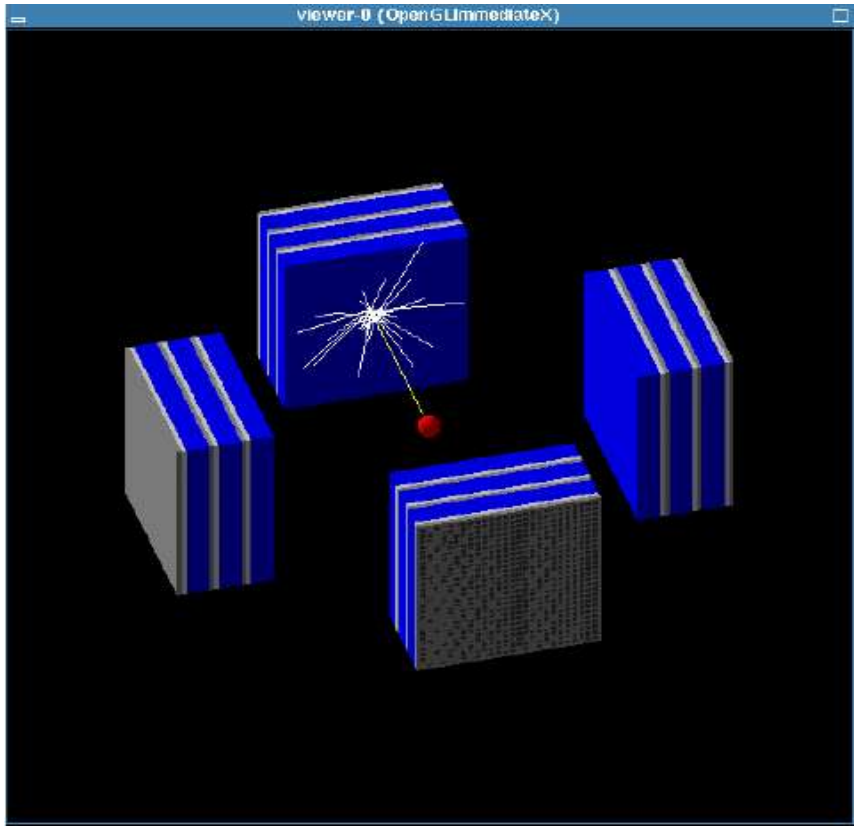


Simulation Set up



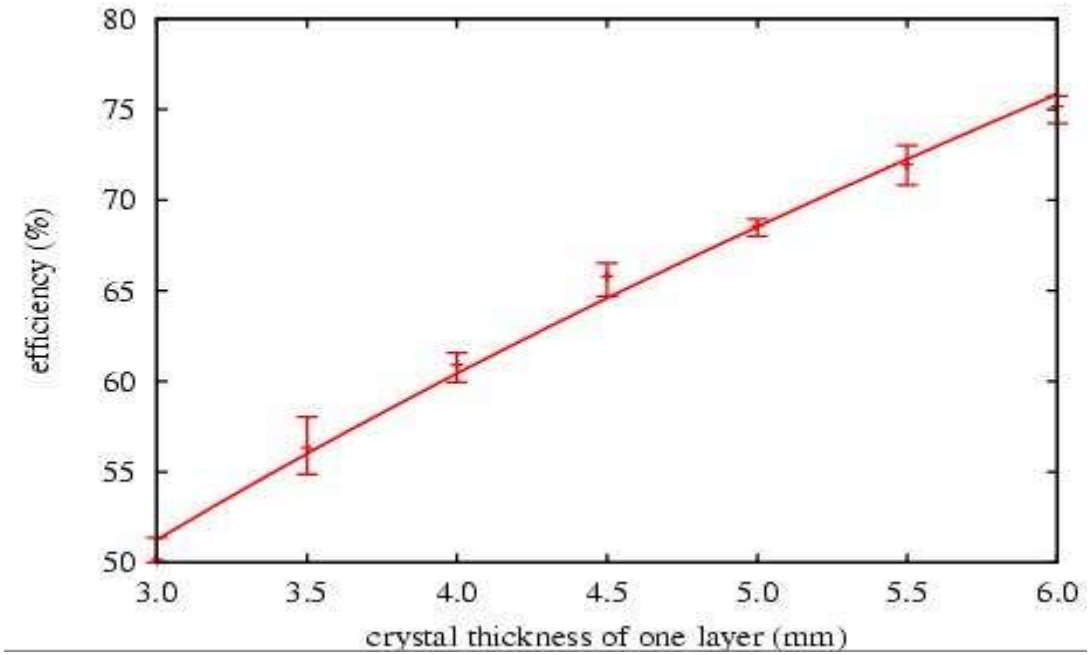






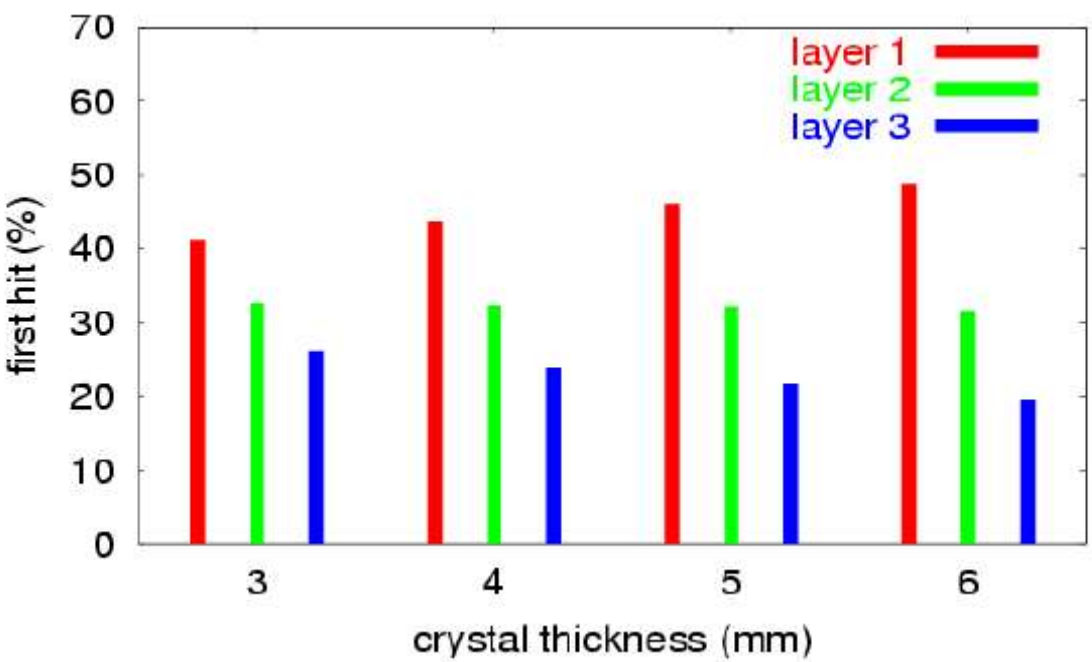
- Simulations made with GEANT 4
  - parallel beam of 511keV gammas
  - 50keV threshold
  - scintillation photons generated
  - QE used averages 2.5% over LSO emissions
  - SiPMs: 1x1mm^2 on a 1.5mm pitch

Efficiency, Backscatter, Count rate



Efficiency -

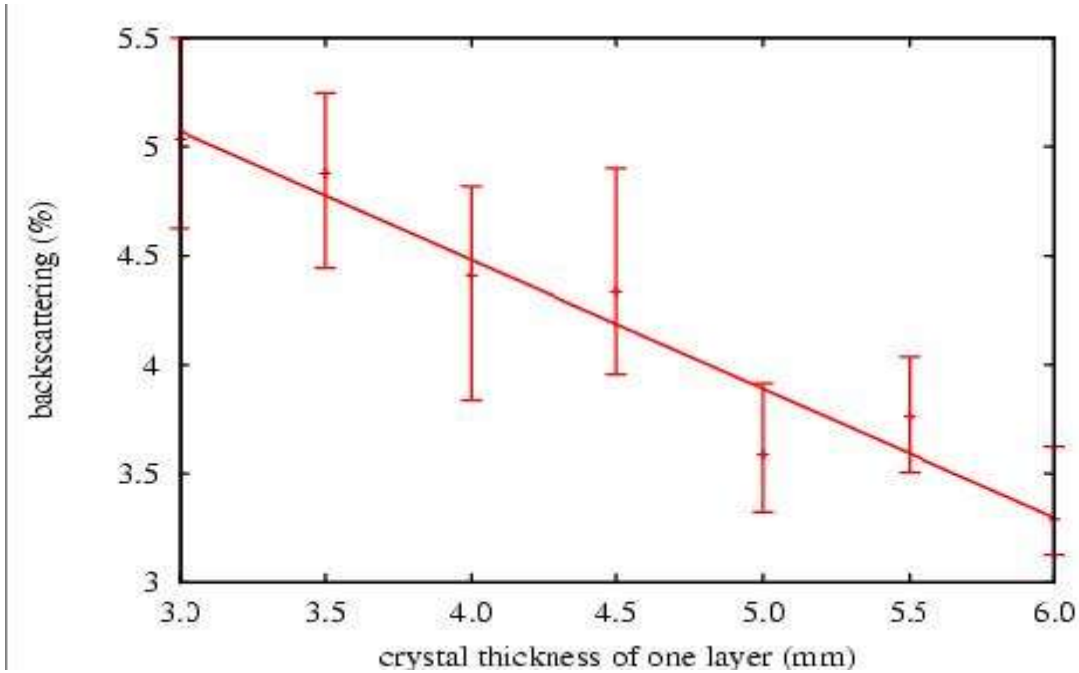
- Percentage of interactions depositing > 50keV in a single layer
- Simply a function of thickness

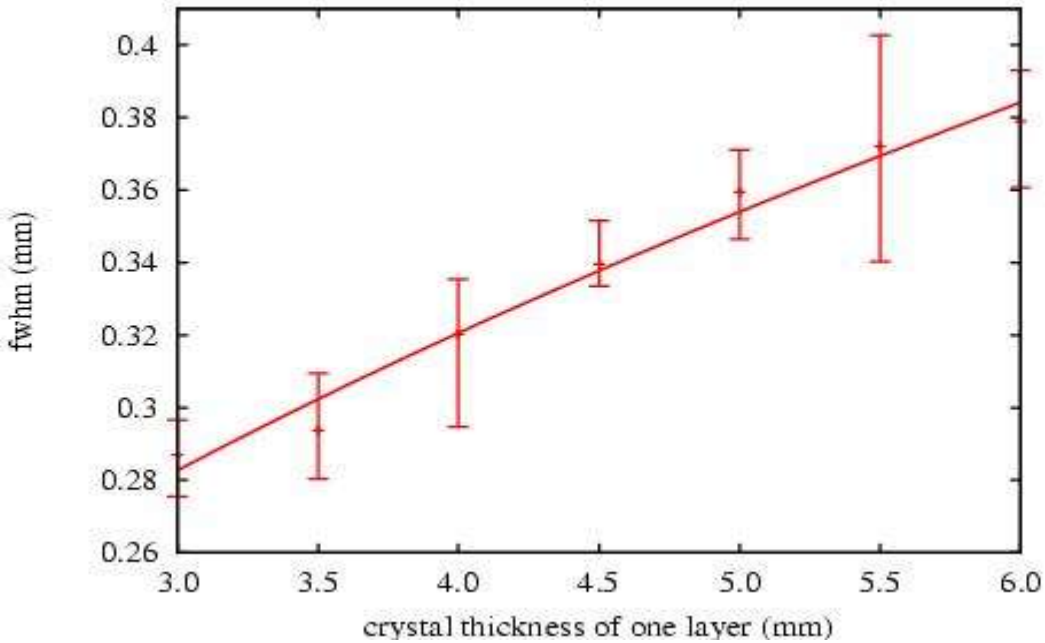


Counts -

- 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%.

Resolution -

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

Parallax contribution

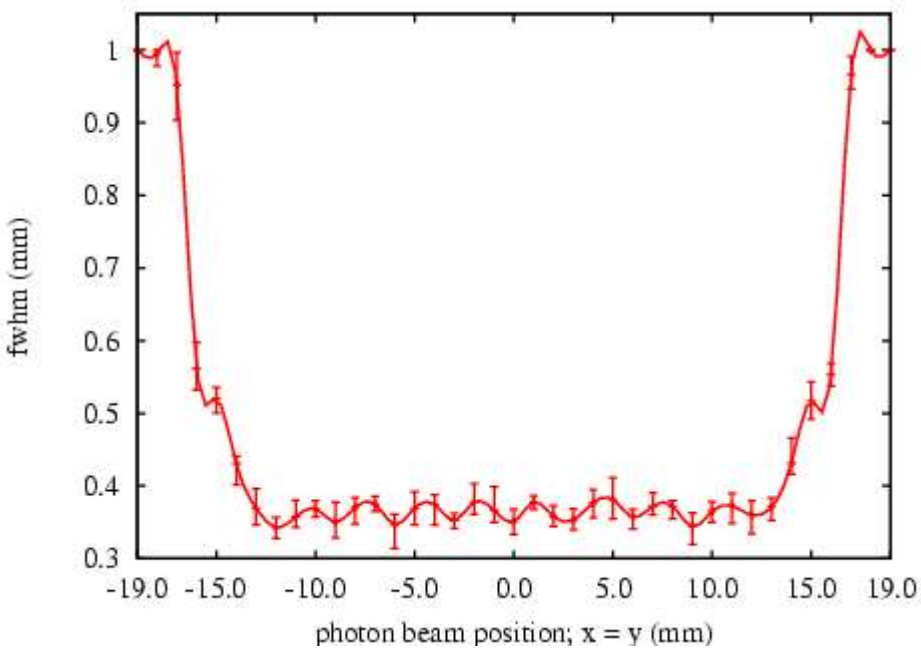
- A thick (5mm) crystal was seen to give good sensitivity and only a small decrease in the FWHM resolution.
- But implies worse DOI resolution which degrades the intrinsic resolution.

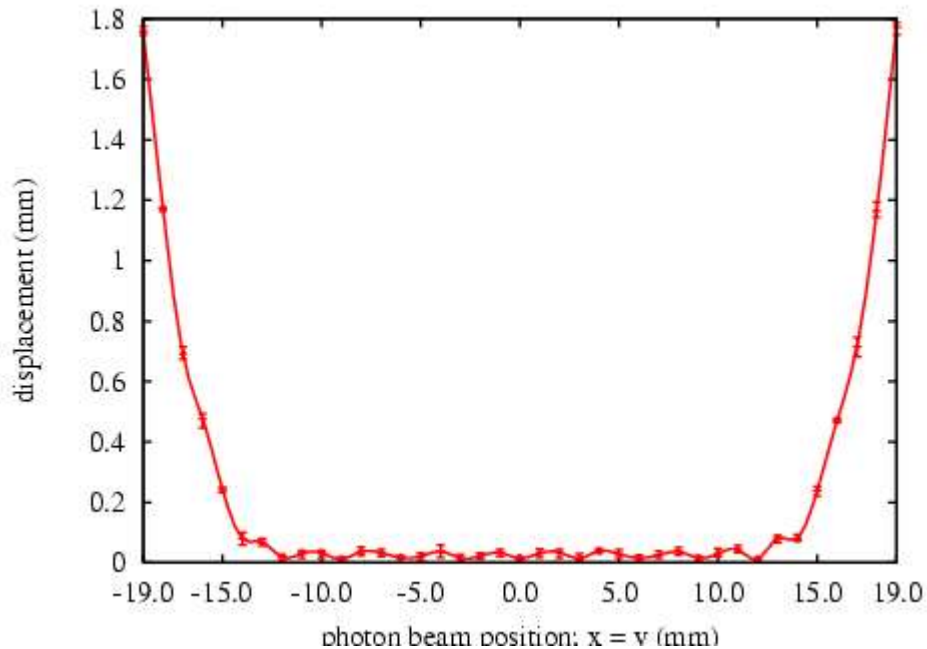
Using a geometrical argument, with skew rays at the maximum of 22°, the contribution to the error in x or y, from the uncertainty in z is ±1mm.

- Additional contributions to the image resolution in PET are
  - Positron range (0.5mm)
  - Acolinearity (< 0.3mm)
  - DOI (1mm?)
  - Intrinsic camera resolution (< 0.5mm)

Edge distortions

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





Summary

- SiPM – Possible replacement for PMT?
- A simple gamma camera design can yield both high spatial resolution and high sensitivity
- Even with poor QE there a high spatial resolution is achieved
- Further simulations for full optimisation

- QE optimised for blue-green
- Matrices development

Future

Many rapid developments

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

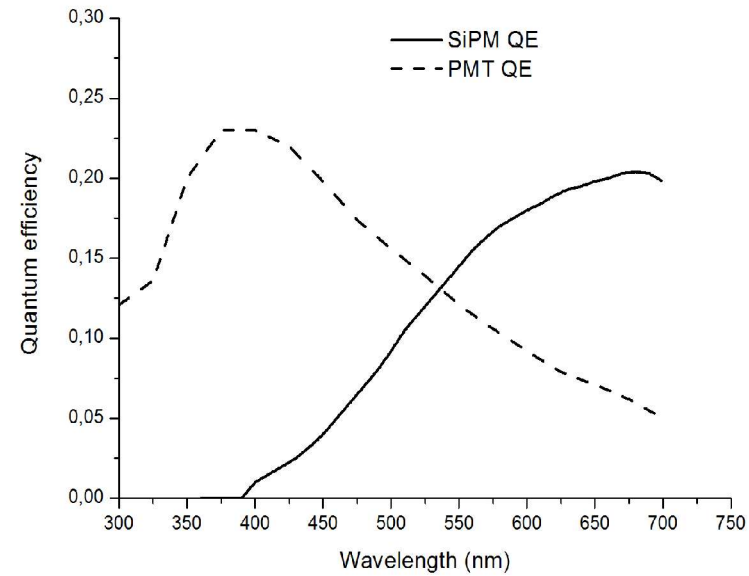
End

Efficiency

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

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

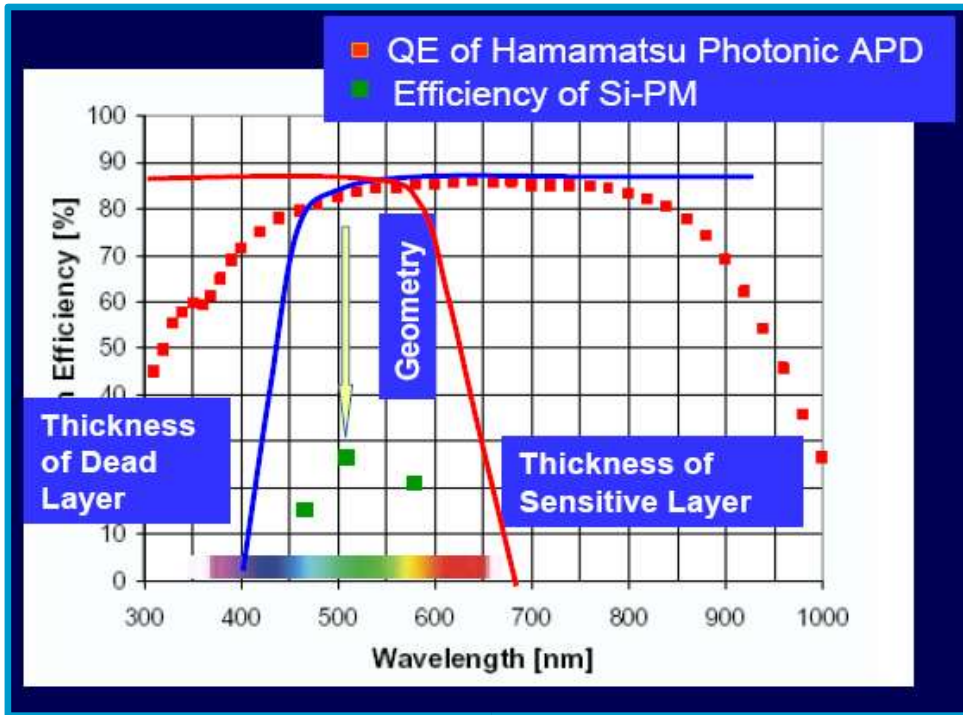
$$\eta = \varepsilon_{geom} \times QE_{tot} = \varepsilon_{geom} \times QE \times \varepsilon_{avalanche}$$



ε_{geom} - geometrical efficiency : $\sum A_{pixel}/A_{total}$

QE_{tot} - total quantum efficiency :

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

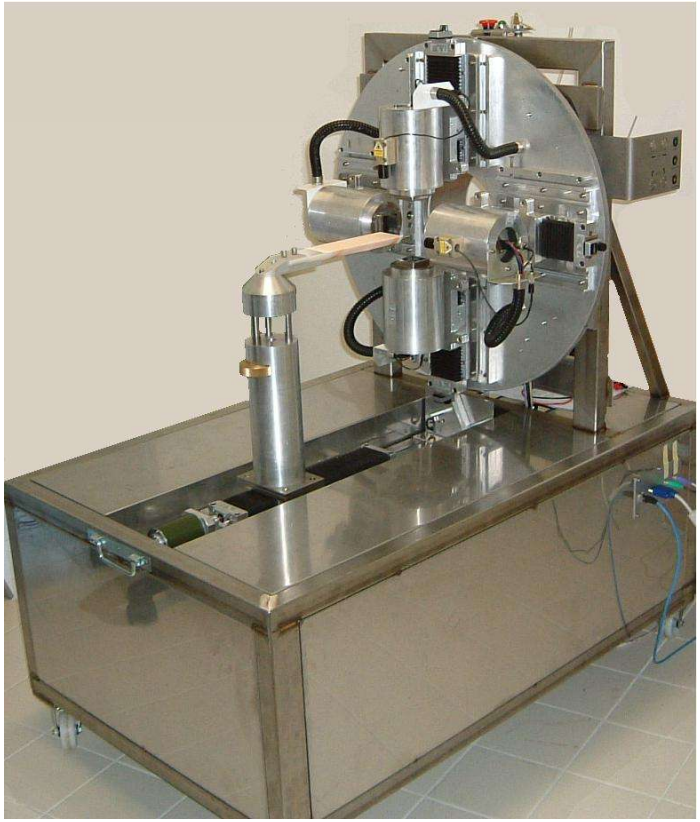


Key element of SiPM efficiency:
geometrical factor = 0.3

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:

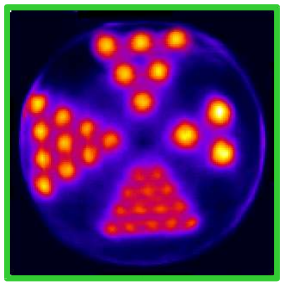
Spatial resolution (iterative algorithm)
Volume: 5.2 mm³ (4.4 mm³) @ CFOV

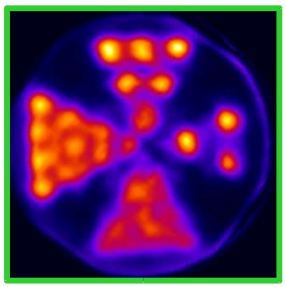
Sensitivity
19 cps/kBq @ CFOV (50-850 keV)

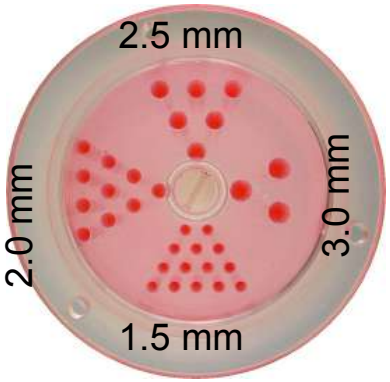
SPECT mode:

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

Sensitivity
30 cps/MBq (constant)







Images of a mini-Dere**nzo** 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 mm³ voxel space (no contrast enhancement).