Frontier Detectors for Frontier Physics
11th Pisa Meeting on advanced detectors
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[redacted]ges and Pitfalls of the Silicon Photomultiplier (SiPM) [redacted]detector for the Next Generation of PET scanners"

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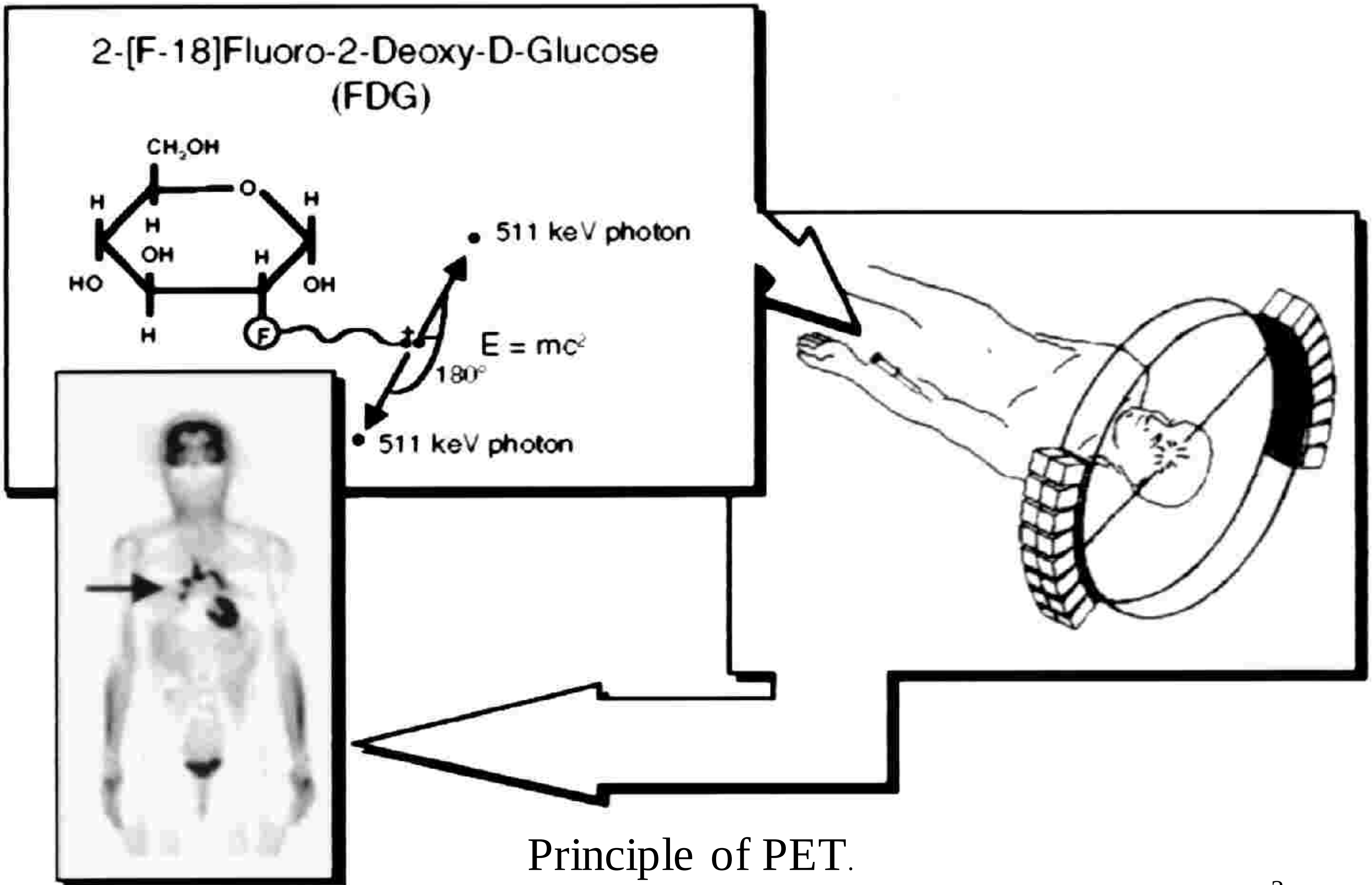
State-of-the-art PET



Principle of PET.

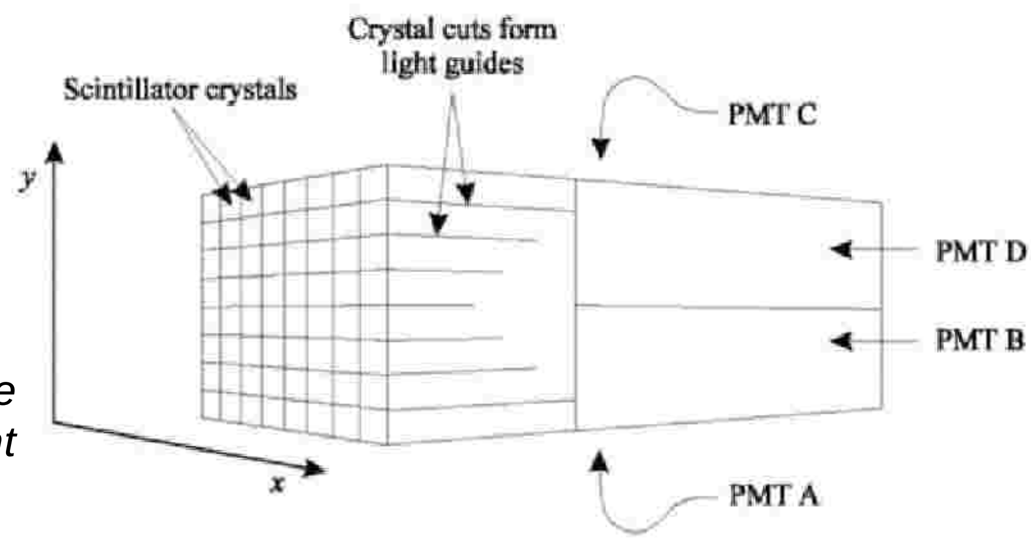
1986 - The block detector

In a block detector, a 2D array of crystals are attached to 4 PMTs. Usually the array will be cut from a single crystal and the cuts filled with light-reflecting material. When a photon is incident on one of the crystals, the resultant light is shared by all 4 PMTs. Information on the position of the detecting crystal may be obtained from the PMT outputs by calculating the following ratios and comparing them to pre-set values:

$$R_x = \frac{A+B}{A+B+C+D}$$

$$R_y = \frac{A+C}{A+B+C+D}$$

where A, B, C and D are the fractional amounts of light detected by each PMT



In 1986 the introduction of the block detector by Mike Casey and Ronald Nutt, changed the world of nuclear imaging. *Almost all dedicated tomographs built since1986 have used some forms of the block detector.*

WHY NEW PHOTODETECTORS?

Increase the EFFICIENCYè Increase SOLID STATE ANGLE

Increase the SPATIAL RESOLUTION

Make use of TIME OF FLIGHT information

PET-MR simultaneous imaging

Detection efficiency

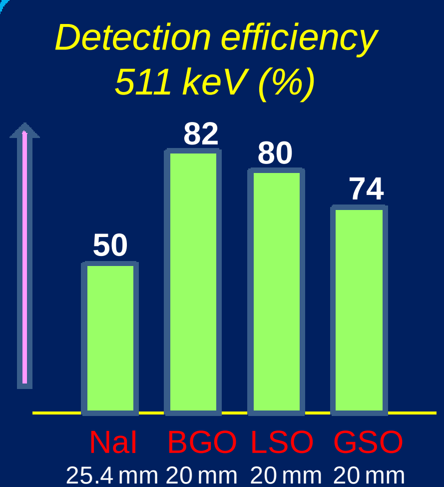
The sensitivity of a PET systems depends mainly on the crystal efficiency and on the system geometry

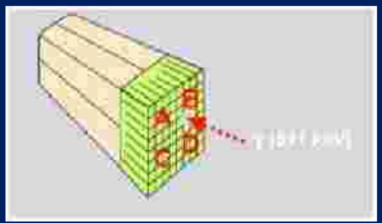
$$h = 100 \cdot \frac{e^{2} \cdot j \cdot \Omega}{4p}$$

ε= crystal detection efficiency

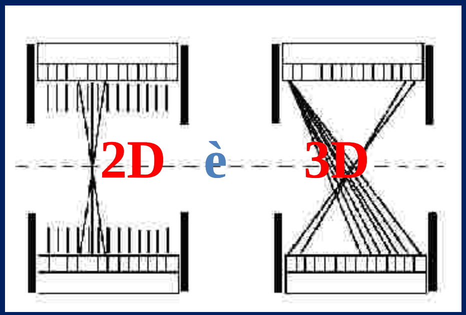
φ= packing fraction

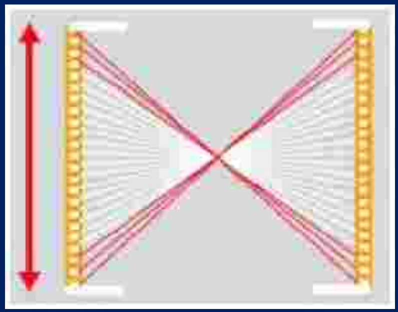
Ω=solid angle





Thicker crystals
20 mm → 30 mm

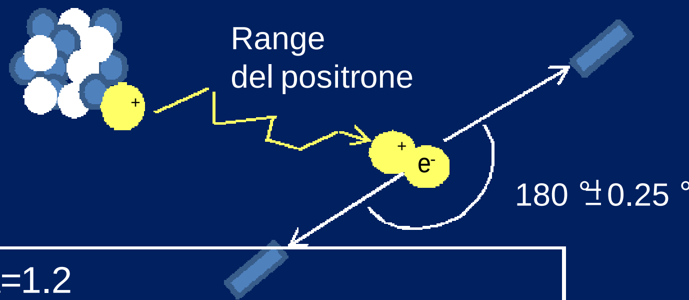




FOV axial extended
16.2 cm → 21.6 cm

Limits to the PET spatial resolution

$$R \approx a\sqrt{R_{range}^2 + R_{180^0}^2 + R_{det}^2 + R_{acc}^2 + R_{DOI}^2}$$



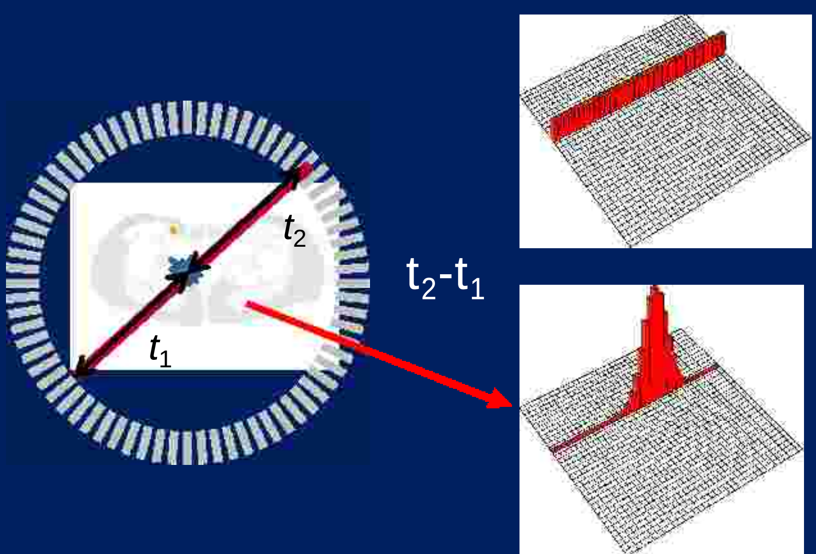
a – degrading factor due to the reconstruction algorithm	1=a=1.2
R_{range} – contribution due to the positron range	R_{range} ~ (FWHM) F-18 0.1-0.2 mm Rb-82 1.27 mm
R_{180} – contribution due to the non- co linearity of the 511 keV photons	R_{180} ~ 0.0022xD D distance between the two coincidence detectors D = 80 cm R_{180} ~ 2 mm (FWHM)
R_{det} contribution due to the detector dimensions	R_{det} ~ d/2 d crystal dimension
R_{acc} – contribution due to the accuracy of the crystal identification algorithm	R_{acc} ~ 1 mm
R_{DOI} – Parallax error contribution due to depth-of-interaction.	R = radius r =distance from center $R_{DOI} = a\frac{r}{\sqrt{r^2+R^2}}$

TOF systems: principle of operation

üTOF-PET systems exploit the time difference between the two emitted photons to better locate the annihilation position.

üThe limit in the annihilation point location is mainly due to the error in the time difference measurement , namely the time resolution **t** of the coincidence system

ü Time resolution is used by the reconstruction algorithm to locate the annihilation point **x** (**x** =c t/2)



PET traditional

The probability for the event to be located along the LOR is uniform

PET Time-of-Flight

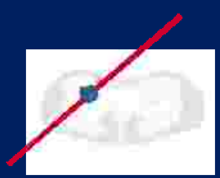
The most likelihood position is in the center of the error distribution

TOF systems: signal to noise ratio

The gain in terms of SNR of the images acquired with TOF-PET systems is proportional to the object dimensions and inversely proportional to the time resolution.

$$SNR_{TOF} \approx \sqrt{\frac{2D}{c\Delta t}} \cdot SNR_{non-TOF}$$

D= diameter of the acquired object
c= light speed
t= time resolution




no TOF **TOF 600 ps** **TOF 300 ps**

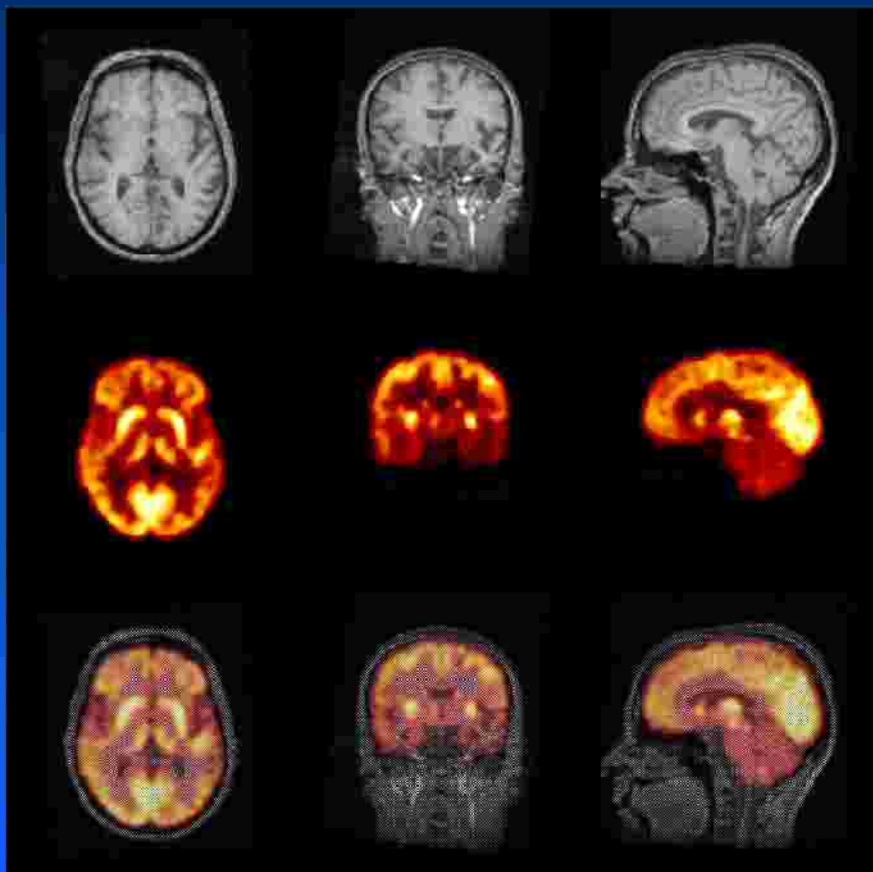
PET +MR

[^{18}F]FDG + MR:
Semantic Dementia

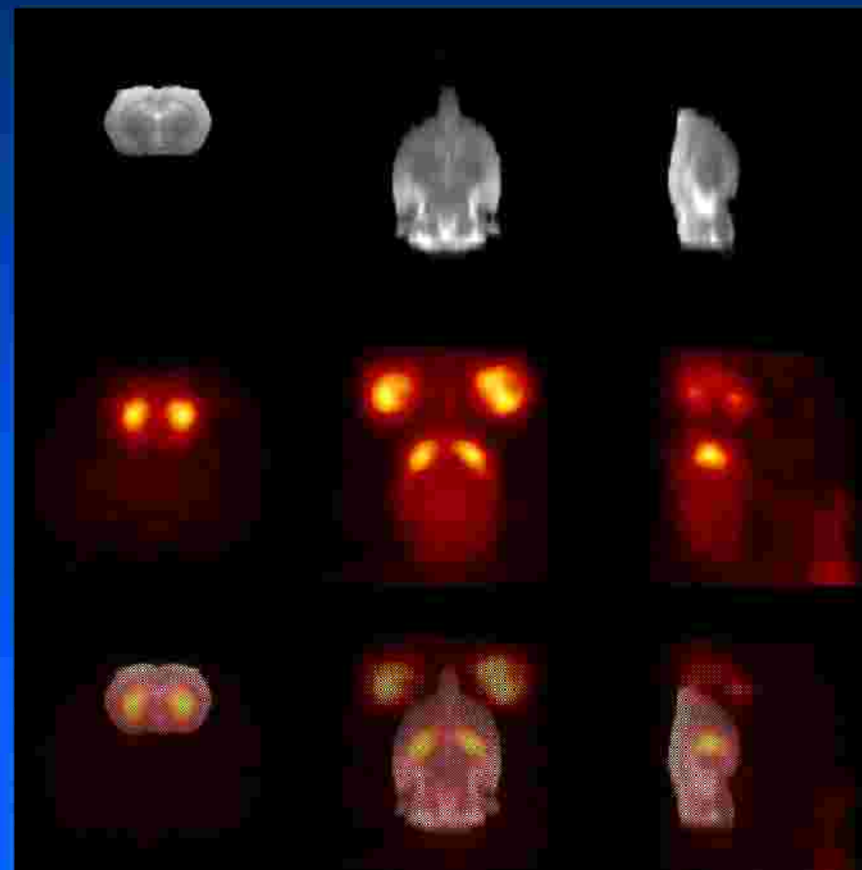
MR

[^{18}F]
FDG

Fused



RAC + MR:
Neuroreceptors study on rat



Technical Challenges in PET/MR

Interference on PET (photomultiplier and electronics)

- Static magnetic field
- Electromagnetic interference from RF and gradients

Interference on MR (homogeneity and gradients)

- Electromagnetic radiation from PET electronics
- Maintaining magnetic field homogeneity
- Eddy currents
- Susceptibility artifacts

General Challenges

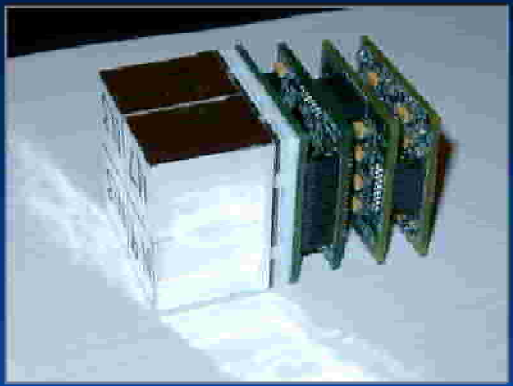
- Space
- Environmental factors (temperature, vibration…)
- Cost

PET attenuation correction via MR data is also a challenge!

Technology for MR/PET

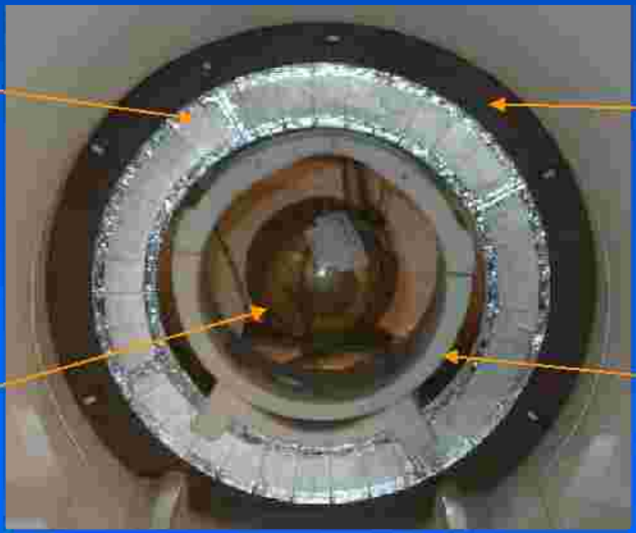
- Solid state devices
  - Avalanche Photodiodes (gain ~ 150)
  - Silicon Photomultiplier (gain ~ 10^6)
  - Less well established as PET detectors
- Can operate in high static field > 7T
- No need to shield devices from both gradients and RF
- Need to shield electronics !

MR-PET Head Insert (with APDs)



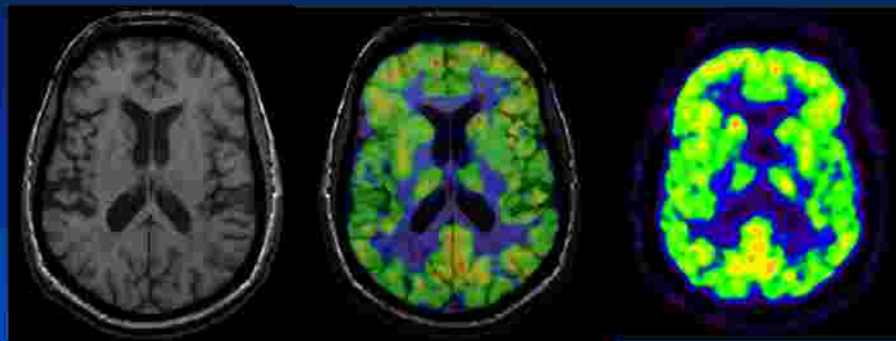
New integrated Detector Block

Prototype PET Head-Insert

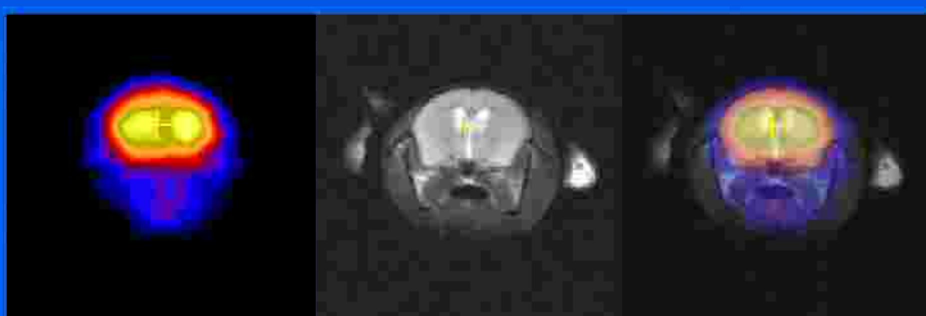


PET/MRI with APDs

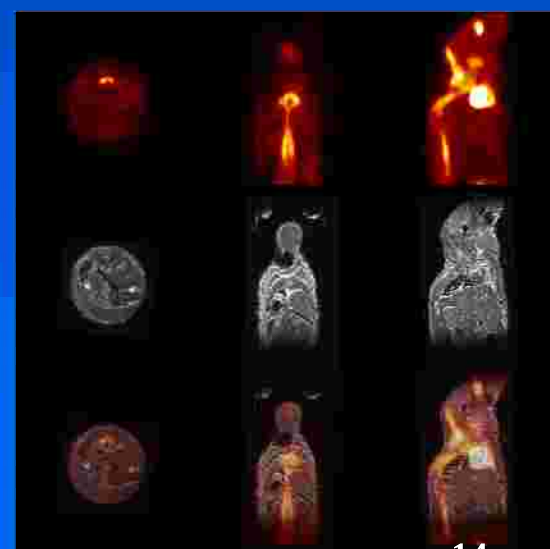
^{18}F-fluorodeoxyglucose - Human



19 x 19 mm crystal block (a): 12 x 12 individual 1.5 x 1.5 x 4.5 mm crystals coupled via a 3 mm thick light guide to a monolithic 3 x 3 APD array (b) (Hamamatsu, Japan)



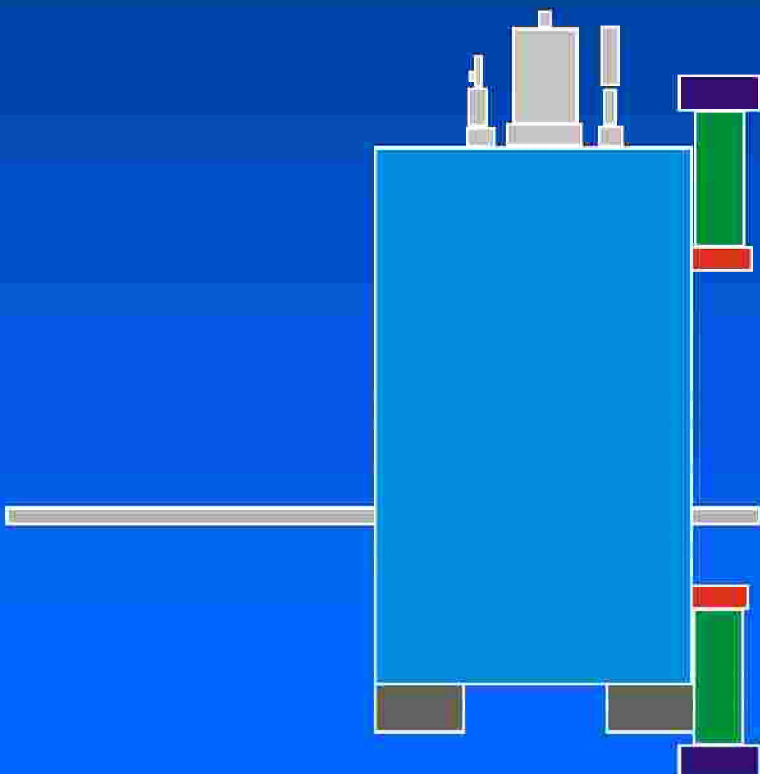
^{11}C-methylphenidate - Mouse

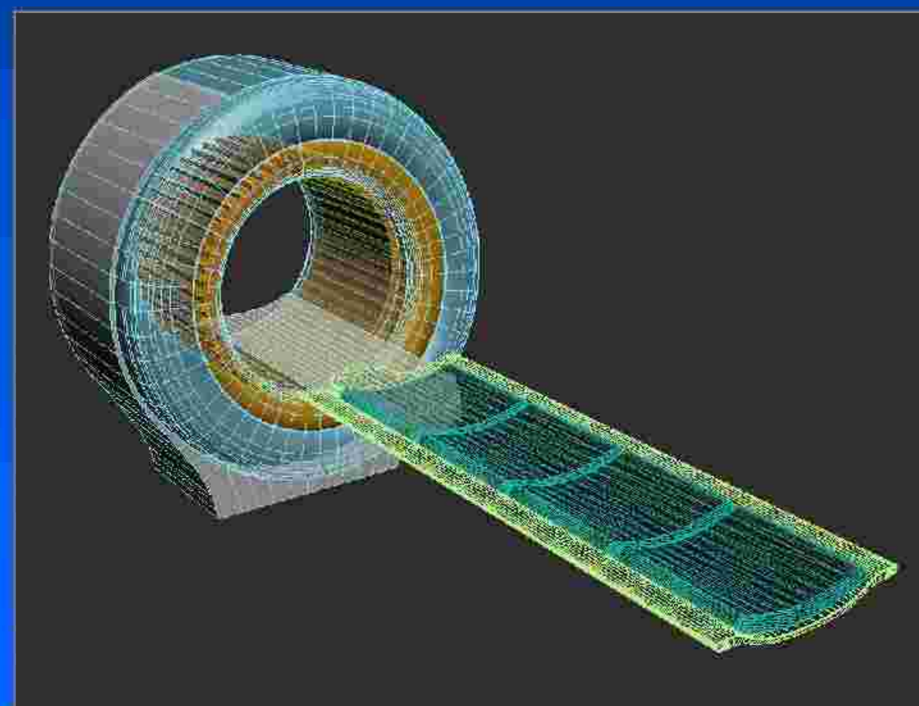


^{18}F-fluorodeoxyglucose - Mouse

Courtesy Berndt Pichler

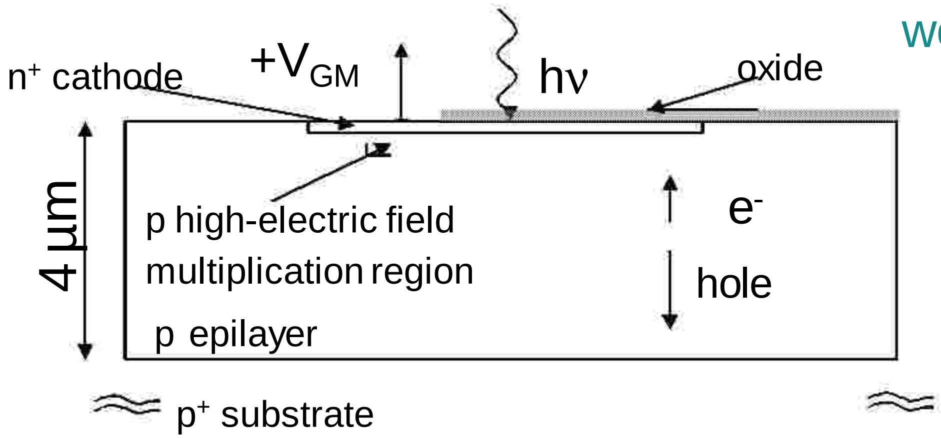
Wholebody MR/PET





Silicon **P**hoto**M**ultiplier = **SiPM**
The Ultimate dream??

SOLID STATE PHOTODETECTOR è



SiPM: **Multicell Avalanche Photodiode** working in limited Geiger mode

- 2D array of microcells: structures in a common bulk.
- Vbias > Vbreakdown: high field in multiplication region
- Microcells work in Geiger mode: the signal is independent of the particle energy
- The SiPM output is the sum of the signals produced in all microcells fired**.**

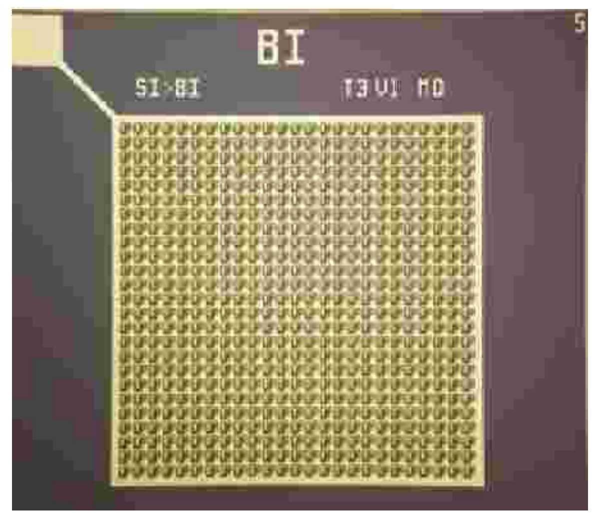
-The photon is absorbed and generates an electron/hole pair

-The electron/hole diffuses or drifts to the high-electric field multiplication region

-The drifted charge undergoes impact ionization and causes an avalanche breakdown.

-Resistor in series to quench the avalanche (limited Geiger mode).

As produced at FBK-irst,Trento, Italyè



à High gain à Low noise à Good proportionality if $N_{photons}$ << N_{cells}

Results: characterization

Collaboration with FBK- irst (Trento, Italy), that has been developing SiPMs since 2005:

First detectors - Single SiPMs (2006)

First matrices 2x2 (2007)

First matrices 4x4 (2008)

First matrices 8x8 (2009)

Breakdown voltage V_{BD} ~ 30V, **very good uniformity.**

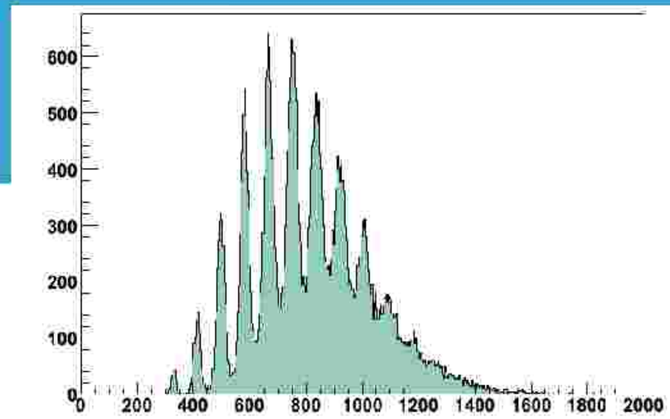
Single photoelectron **spectrum: well resolved peaks.**

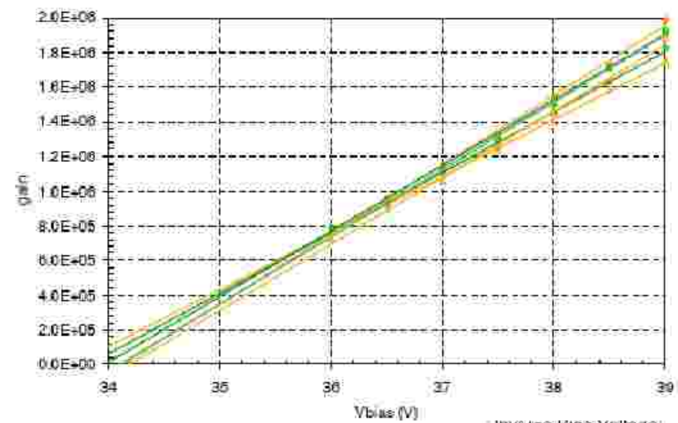
Gain: ~10^6

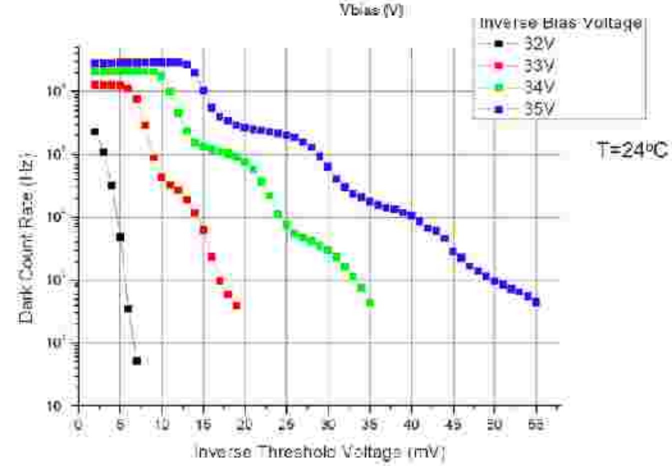
- Linear for a few volts over V_{BD}.
- Related to the recharge of the diode capacitance CD from V_{BD} to V_{BIAS} during the avalanche quenching. $G=(V_{BIAS}-V_{BD}) \times C_D/q$

Dark rate:

- 1-3 MHz at 1-2 photoelectron (p.e.) level, ~kHz at 3-4 p.e (room temperature).
- Not a concern for PET applications.







Results: intrinsic timing

Intrinsic timing measured at s.p.e level:
60 ps (s) for blue light at 4V overvoltage.

SiPM illuminated with a pulsed laser with 60 fs pulse width and 12.34 ns period, with less than 100 fs jitter.

Two wavelengths measured:

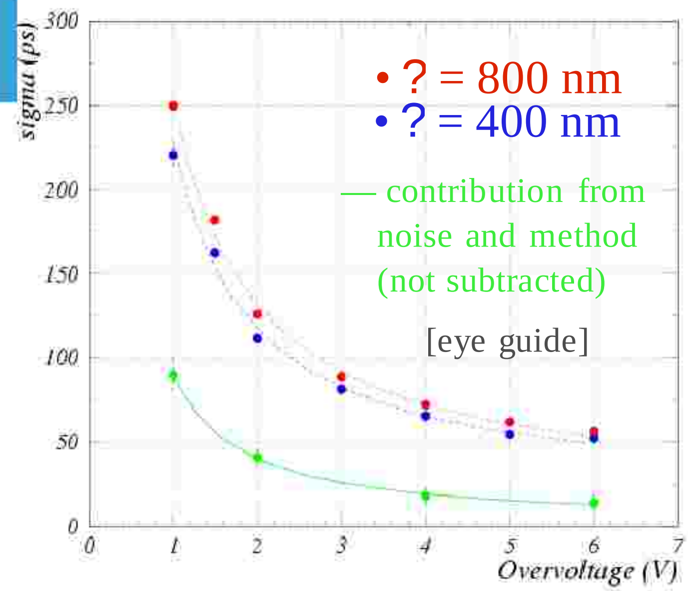
λ = 400 ± 7 nm and = 800 ± 15 nm.

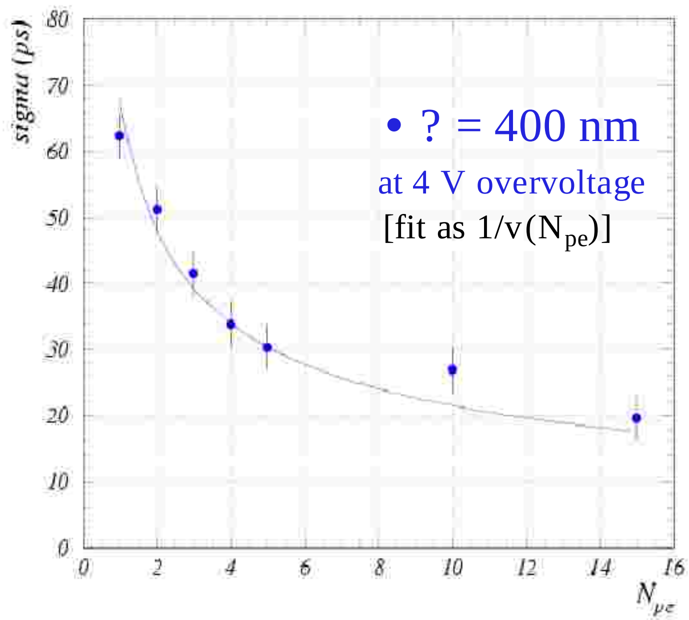
Time difference between contiguous pulses is determined.

The time resolution increases with the number of photoelectrons as

1/v(Npe)è 20 ps at 15 photoelectrons.

[G. Collazuol et al., VCI 2007, NIM A 2007, A581, 461-464]

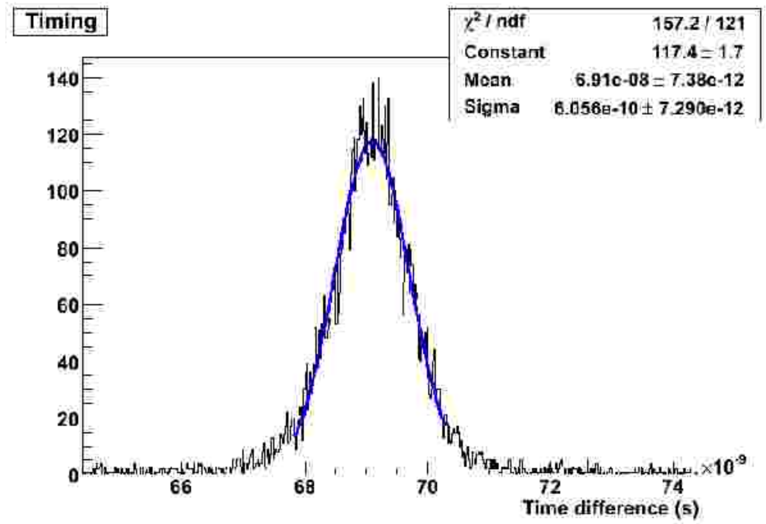




Results: coincidence timing (TOF)

Coincidence measurement with two LSO crystals (1x1x10 mm^3) coupled to two SiPMs **{From Theory: Post and Schiff. Phys. Rev. 80 (1950)1113.}**

$$\sigma \sim \frac{\sqrt{Q}\,\tau}{< N >}$$



Where:

<N> = average number of photons: ~ 100 photons at the photopeak

Q = Trigger level: ~1 photoelectron.

τ = Decay time of the scintillator

For two scintillators in coincidence expected : => √2s ~ 630 ps .

Measured => ~ 600 ps sigma.

Measurements in agreement with what we expect!!

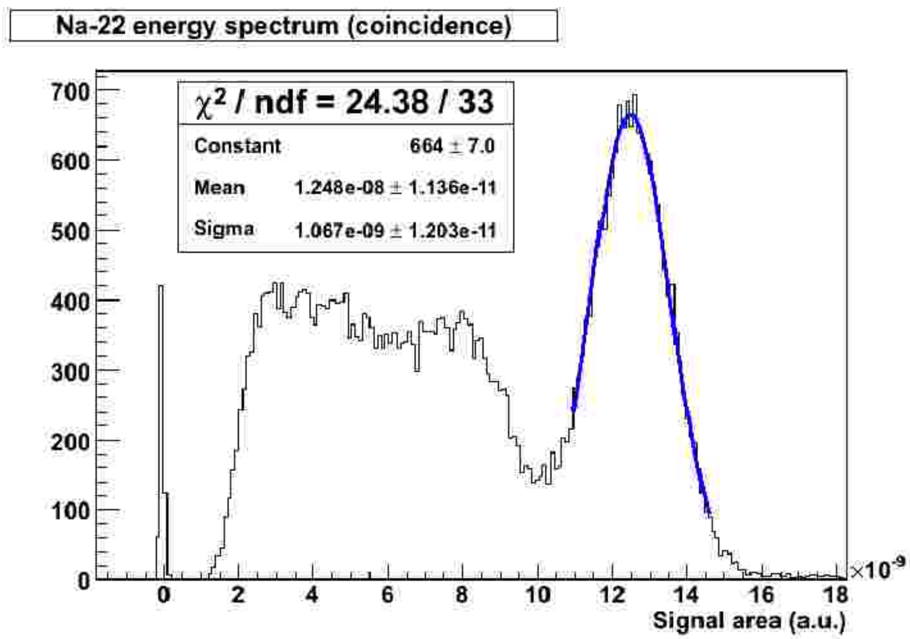
[G.Llosa,et al., IEEE Trans. Nucl. Sci. 2008, 55(3), 877-881.

Results: energy resolution (DE/E)

Setup:

- 2 LSO [1mm x 1mm x 10mm] crystals coupled to 2 SiPMs
- Home made amplifier board.
- Time coincidence of signals.
- VME QDC for DAQ.
- ^{22}Na source.

Energy resolution in coincidence: 20% FWHM.
(best result: 17.5 %)



[G.Llosa et al, IEEE Trans. Nucl. Sci. 2008, 55(3), 877-881.]

Results: tests of SIPM in MR system (MRI)

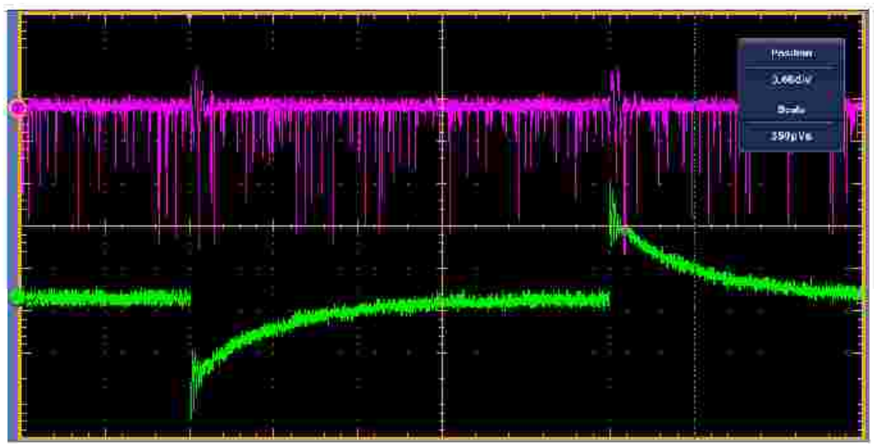
in collaboration with the Wolfson Brain Imaging Center, Cambridge, UK

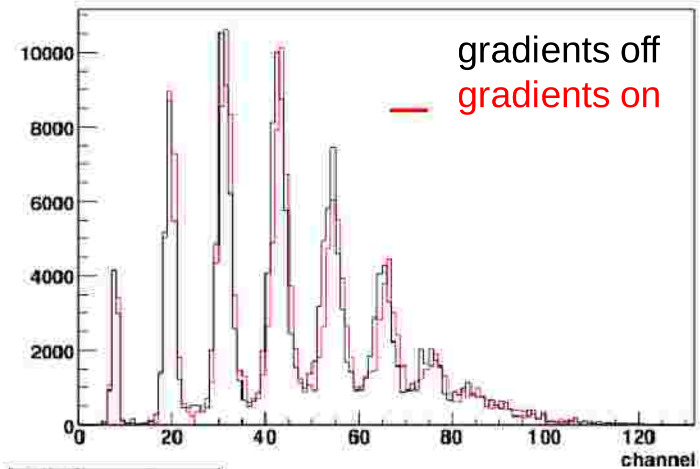
S.p.e and ^{22}Na energy spectra acquired with gradients off (black line) and on (red line).

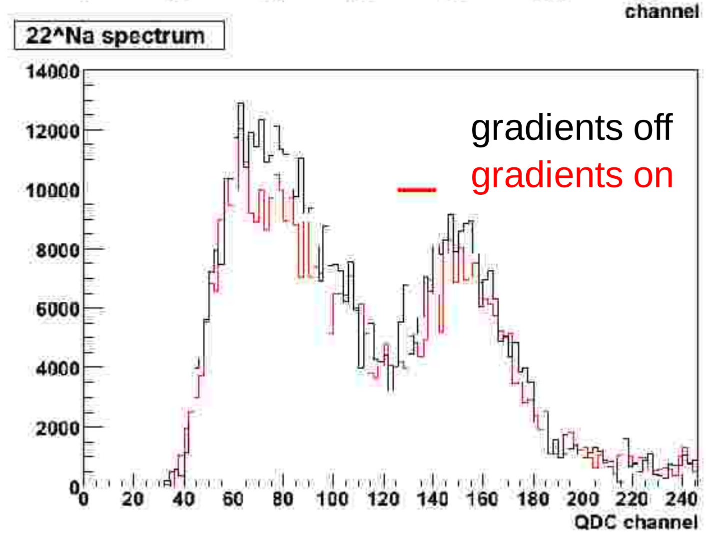
No real difference is appreciated in the data.

Differences in photopeak position is due to temperature changes in the magnet **(apparent change in gain due to changes in breakdown voltage).**

Pickup in baseline when switching on/off







[R.C.Hawkes,et al. 2007 IEEE NSS-MIC, Honolulu, USA, October 28-November 3, 2007: M18-118.]

Results: New detectors (May 2007)

Different geometry,size,microcell size and GF.

40x40 m^2 => GF 44%

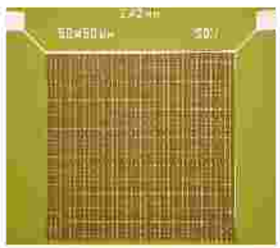
50x50 m^2 => GF 50%

100x100 m^2 => GF 76%

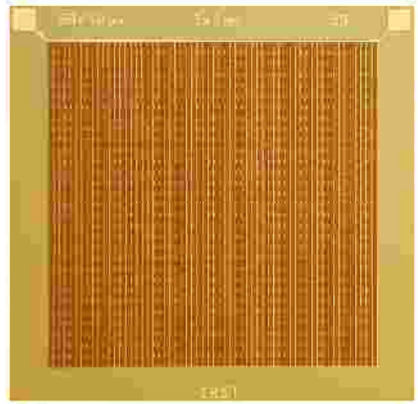
circular

1mm ϕ

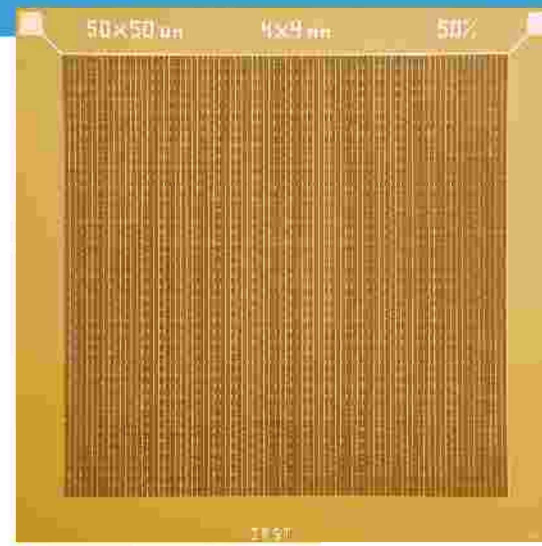
1x1mm^2



2x2mm^2



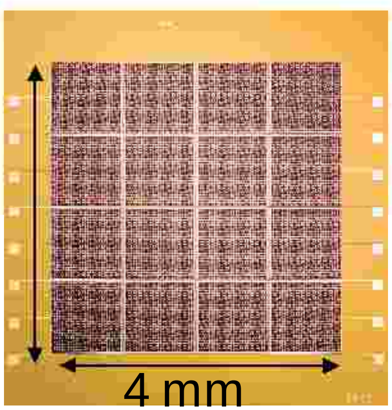
3x3mm^2 (3600 cells)



4x4mm^2 (6400 cells)

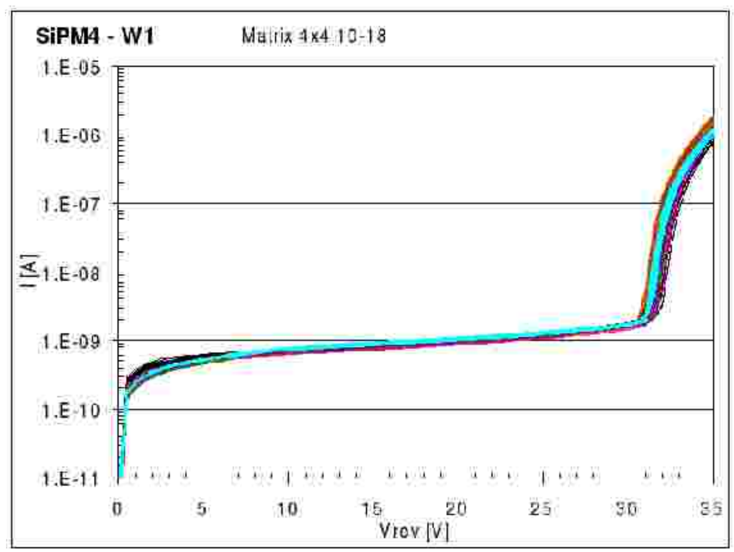
Matrices 16 elements (4x4)

4 mm



IV CURVES OF 9 MATRICES.

VERY UNIFORM BREAKDOWN POINT

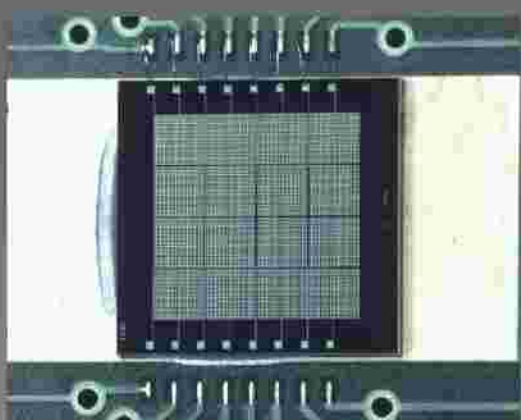


[C.Piemonte et al, Il Nuovo Cimento C, 2007,30(5),473-482]

SiPM 4x4 matrices from FBK-irst

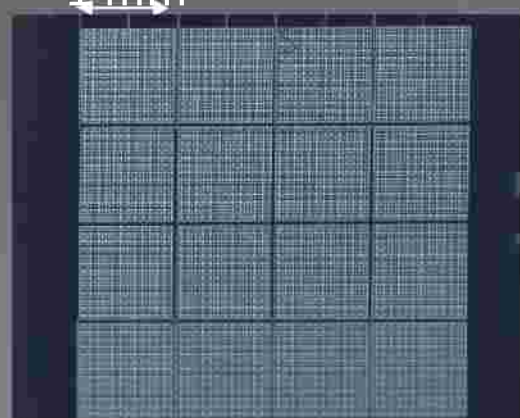
Composed of 16 (4x4) pixel elements in a common substrate
1 mm pixels in 1.06 mm pitch

- Structure: n^+-p-   -p^+ optimized for blue light: Shallow n^+ layer + specific antireflective coating.
- Each pixel: 625 (25 x 25) microcells, 40 m x 40 m size.
- Polysilicon quenching resistor.
- Fill factor 44%.

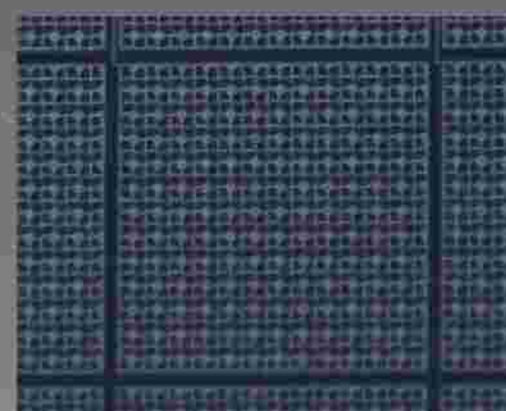


Bonded SiPM array

1 mm



SiPM array

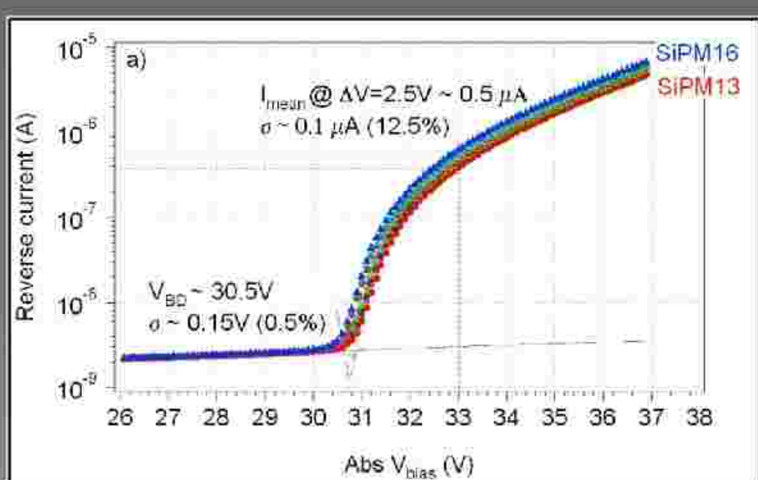


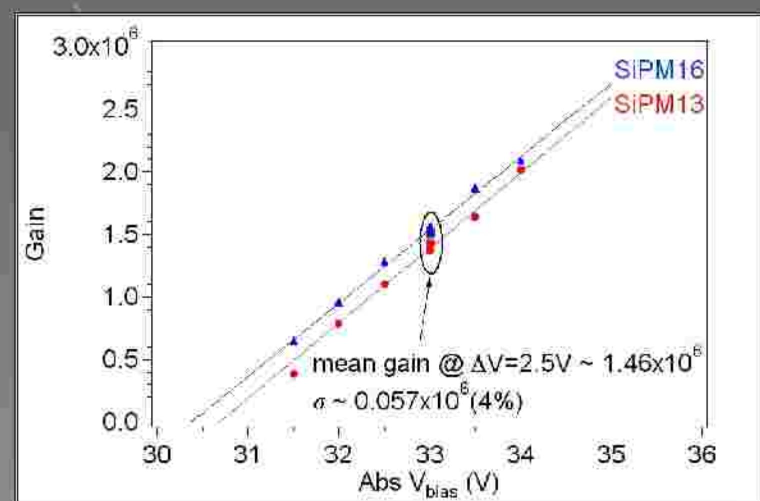
SiPM pixel

4x4 Matrices Characterization

- The full characterization of the first production was performed at LAL, Orsay.
- Excellent uniformity.
  - Breakdown voltage 30.5V ; $_{var}$= 0.5%
  - Gain @33V 1.46x10^6 $_{var}$ 4%
- Mean dark rate @33V (V=2.5V): 1.98 MHz
- PDE @ 33V 8-10% from 420 to 680 nm wavelength.

Expected PDE >15% for the results shown at V=4V

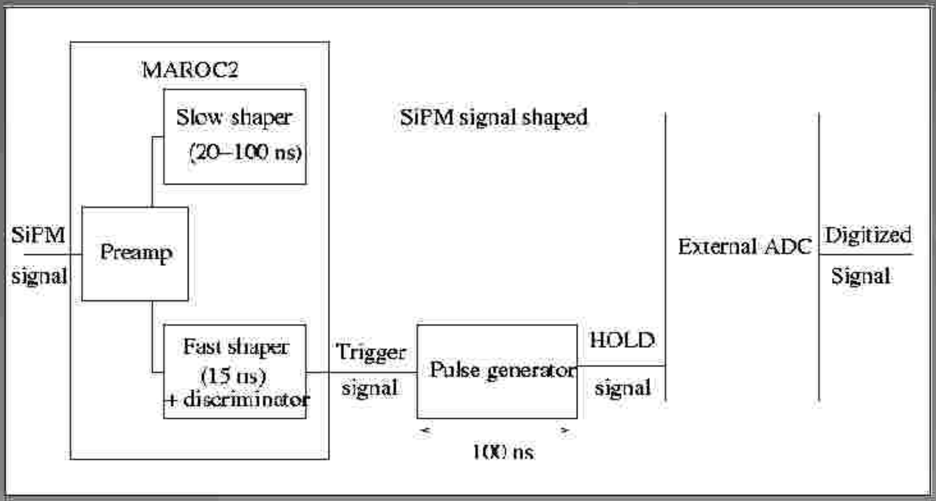




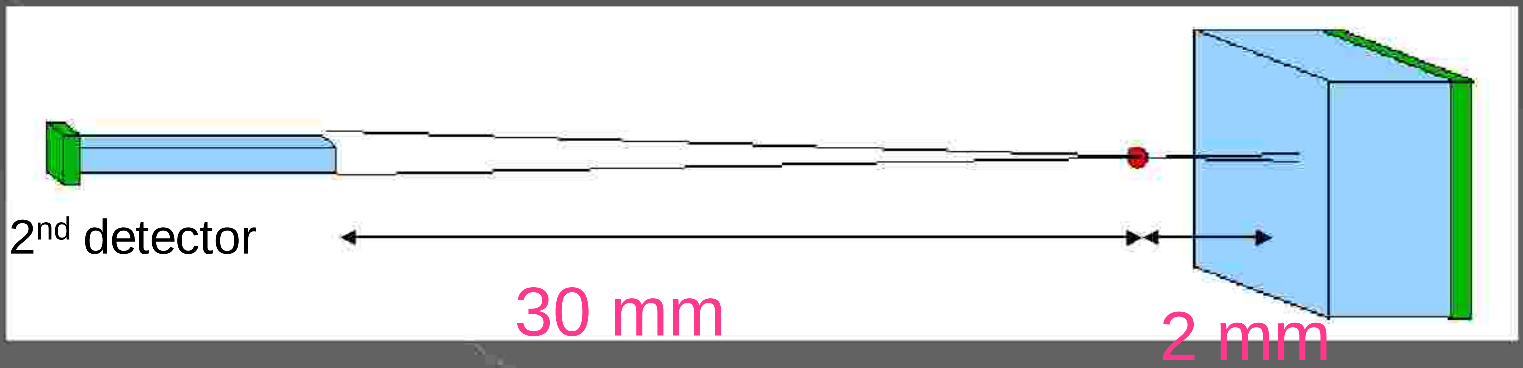
N. Dinu et al., Pixel 2008 workshop,,Fermilab, Septemb 2008.

Readout: MAROC2 ASIC

- Developed at Laboratoire de l'Accelerateur Lineaire, Orsay.
- 64 channels
- low noise preamplifier with variable gain (6 bits)
- Slow shaper (~20-150 ns, adjustable)
- Fast shaper (15 ns) + 3 discriminators =>Trigger signal.
- Designed for MAPMT (H8500)– not optimized for SiPMs, but allows us to make the tests satisfactorily.



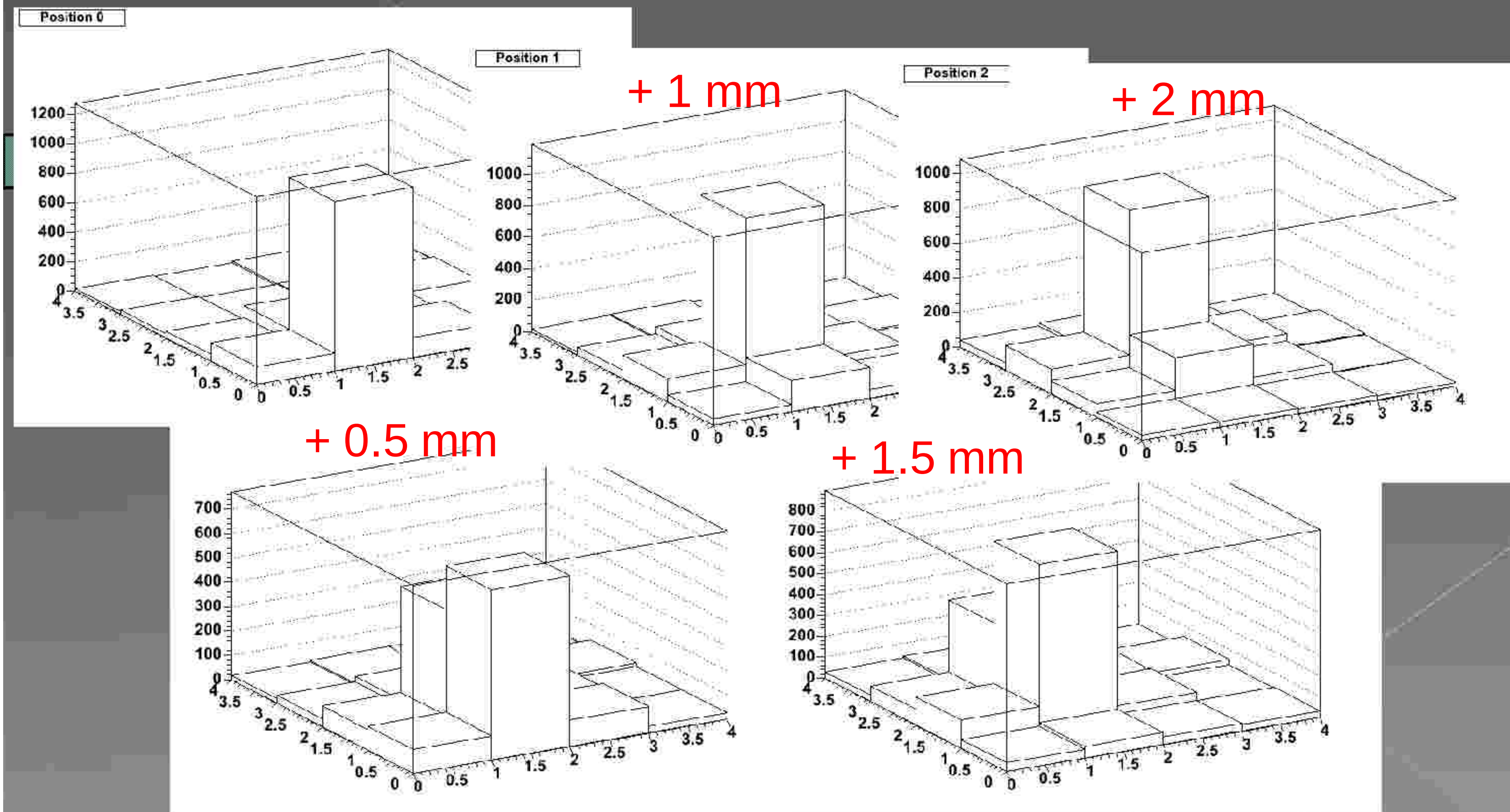
Position determination



- Coincidence with a 2nd detector: 1 mm x 1 mm x 1 cm crystal coupled to a SiPM
- Source close to the matrix, far from 2nd detector
- Move together source and 2nd detector.

Position determination-crystal array

Hit map for different source positions with crystal array



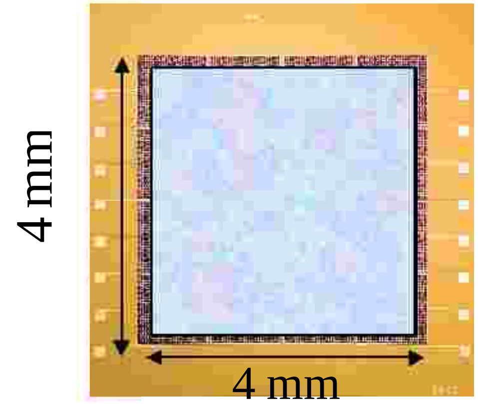
Results with continuous crystals

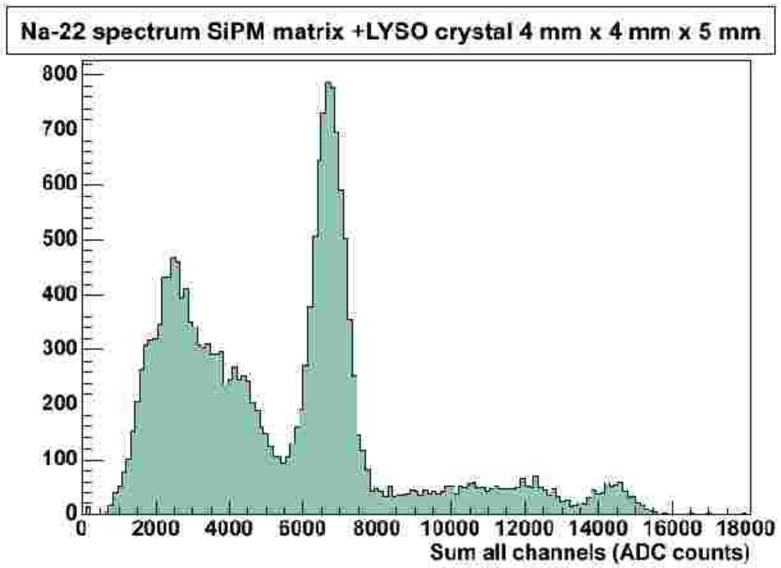
Crystal 4 mm x 4 mm x 5 mm covering the whole 4x4 matrix.

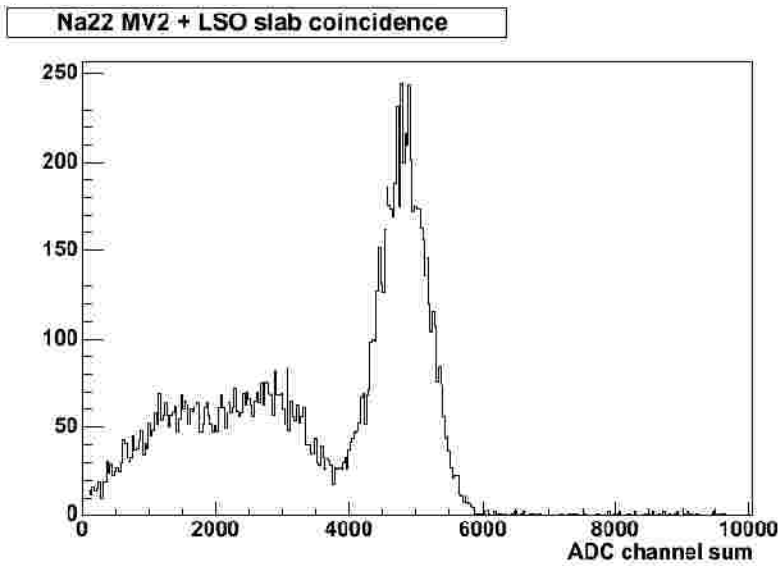
Na-22 spectrum summing signals from all channels.

$V_{over\text{-}br} = 4V$

$\Delta E/E = 16\%$

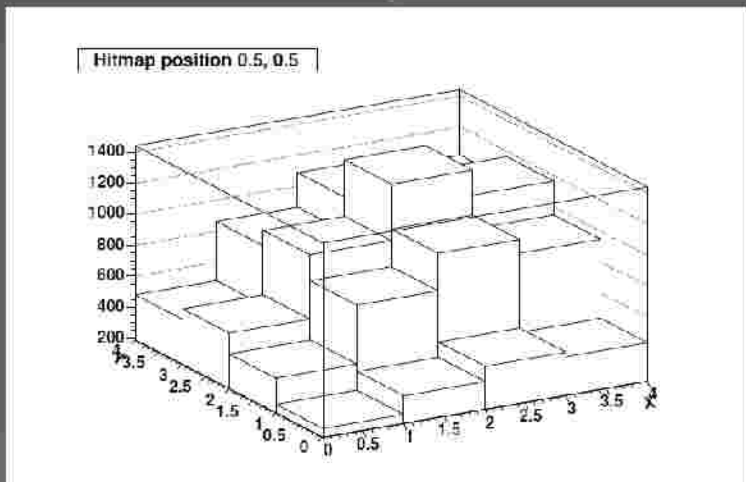


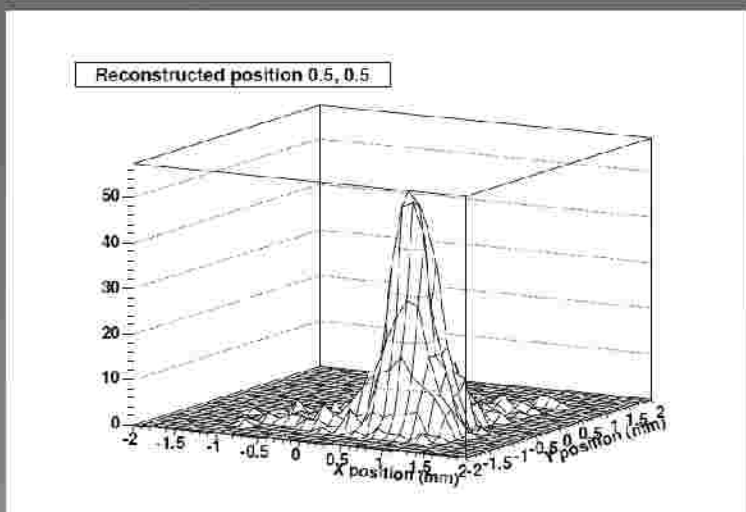




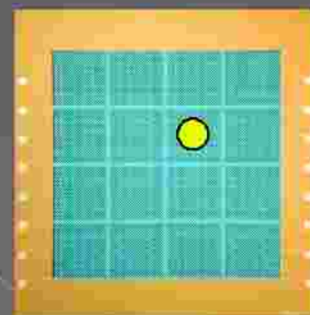
G.Llosa et al., Submitted to IEEE TNS, 2009

Position determination-
-black slab





- Hit map

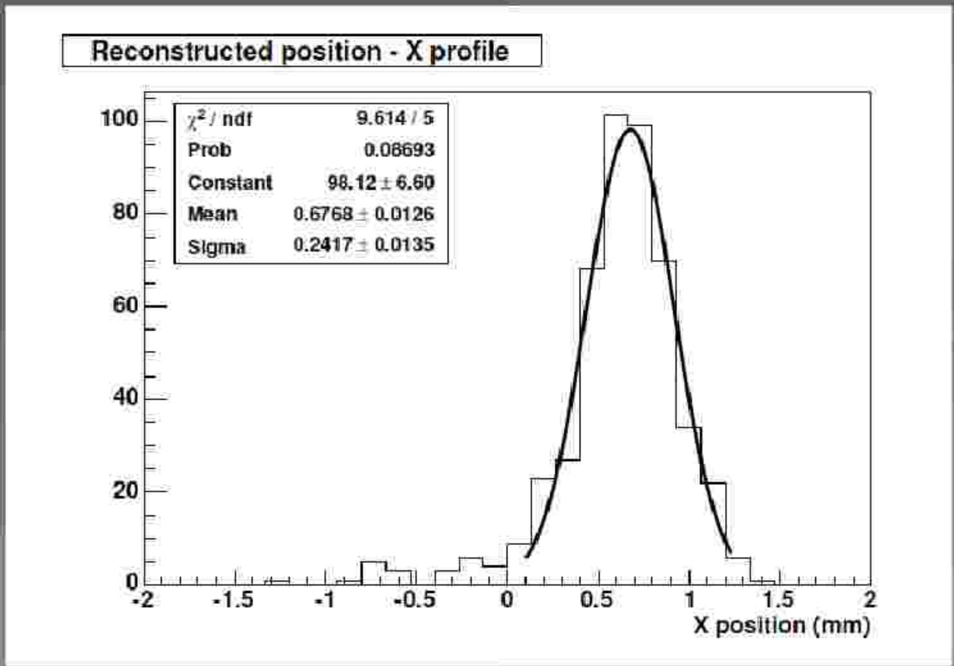


- “center of gravity” Algorithm

$$X = \frac{\sum X_i ADC_i}{\sum ADC_i},$$
$$Y = \frac{\sum Y_i ADC_i}{\sum ADC_i},$$

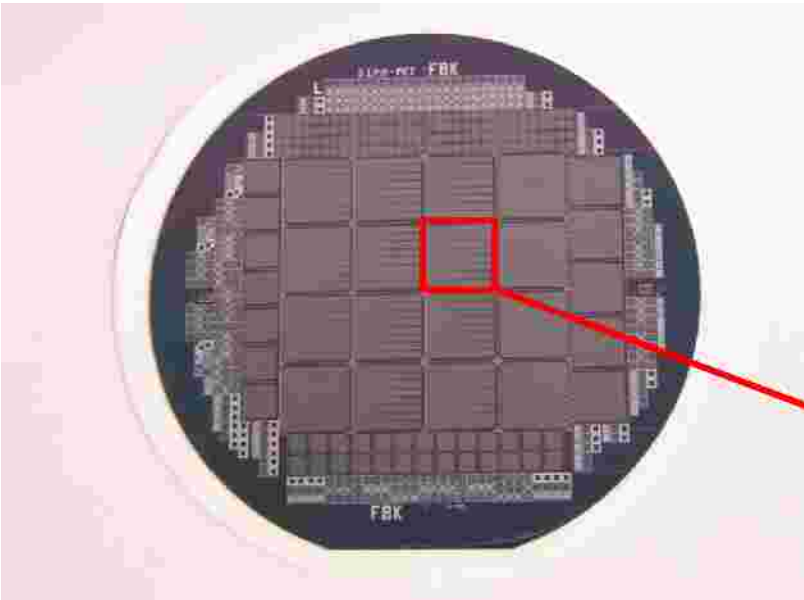
Position determination-black slab

- Matrix + LYSO crystal 4mm x 4mm x 5mm painted black
- Center of gravity algorithm – problems at the edges
- Difficulties due to the small size of the devices
- Intrinsic spatial resolution: 0.57 mm (FWHM) at CFOV

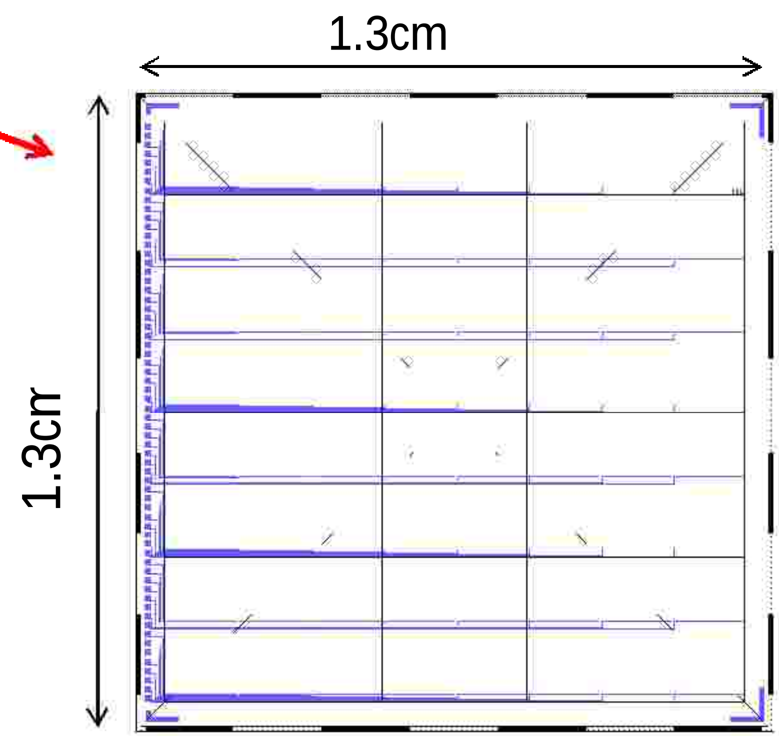


G.Llosa et al., Submitted to IEEE TNS, 2009

Matrices for INFN-DaSiPM2 project (2009)



- 8x8 matrix
- 1.5mm element pitch
- 625 (50 m x 50 m) cells
- read-out on one side



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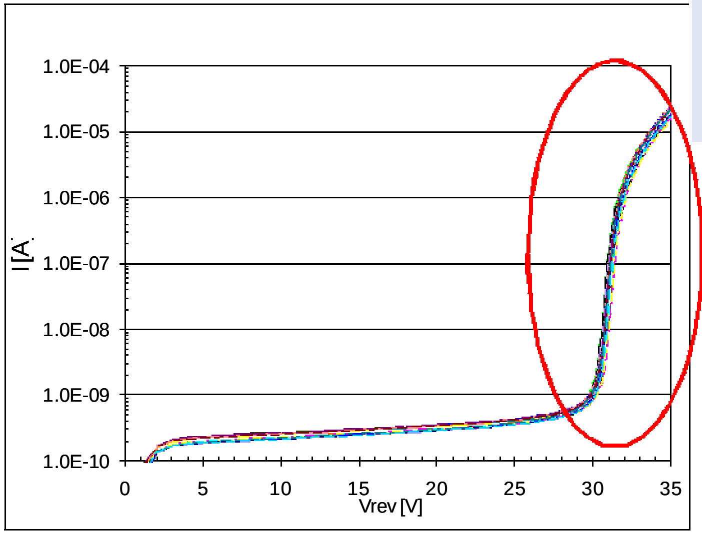
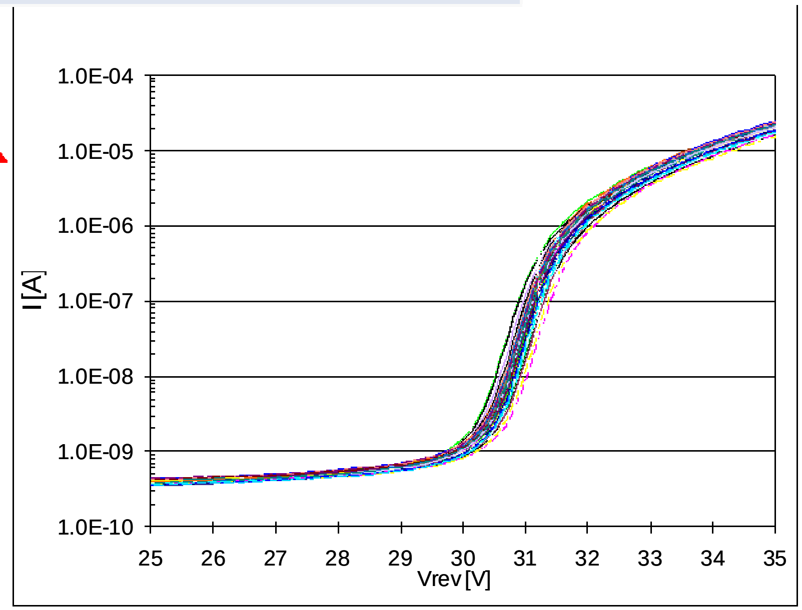
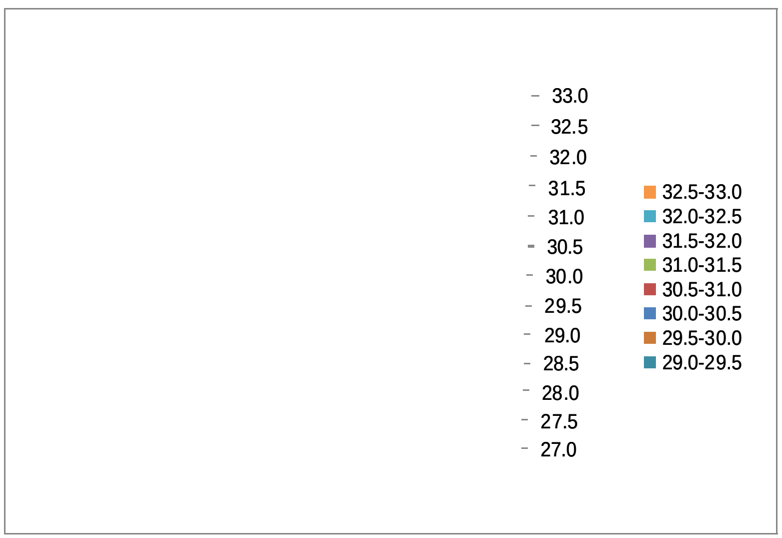
DaSiPM2 8x8 Matrices (2009)



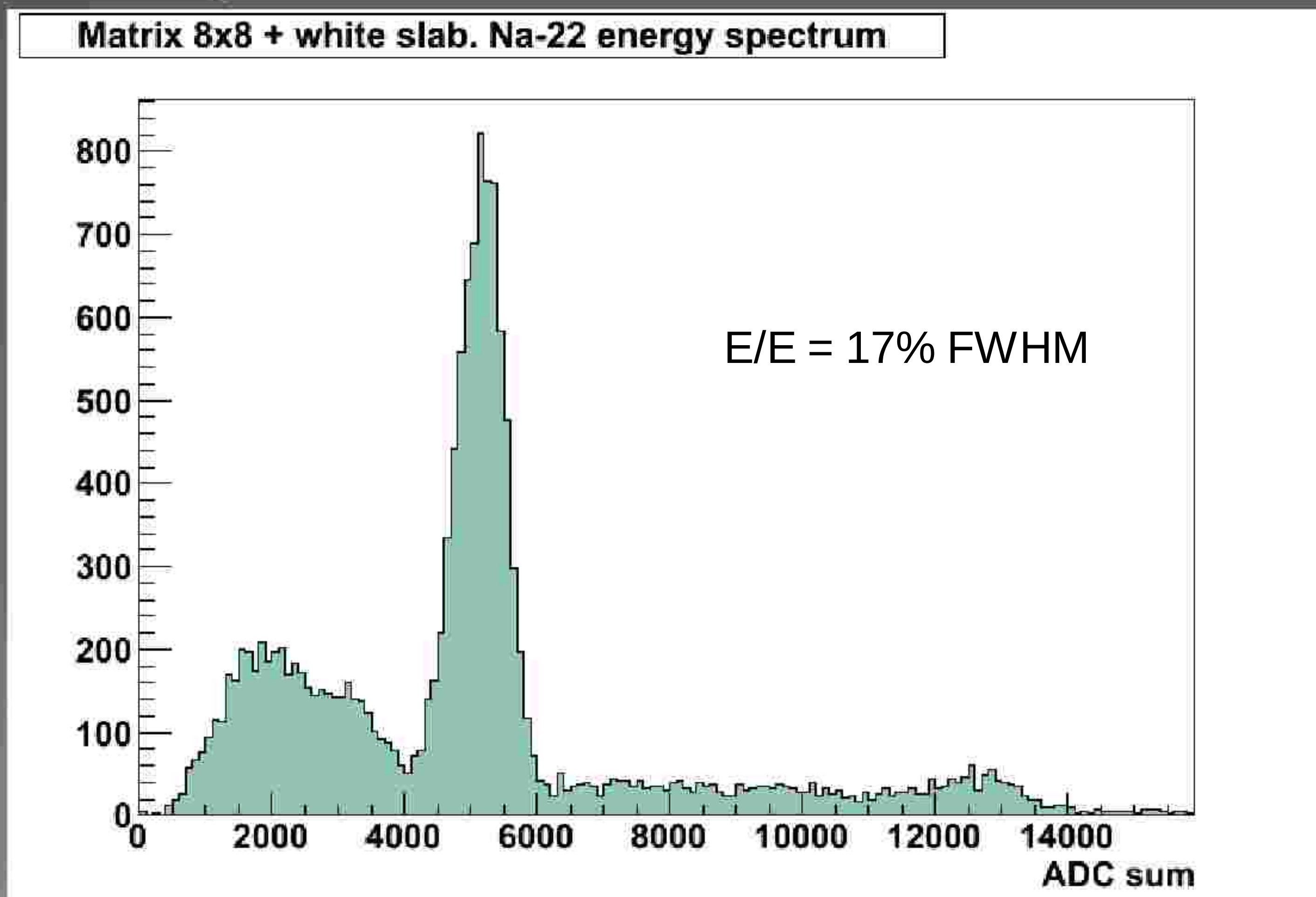
IV plot of the 64 elements of the matrix





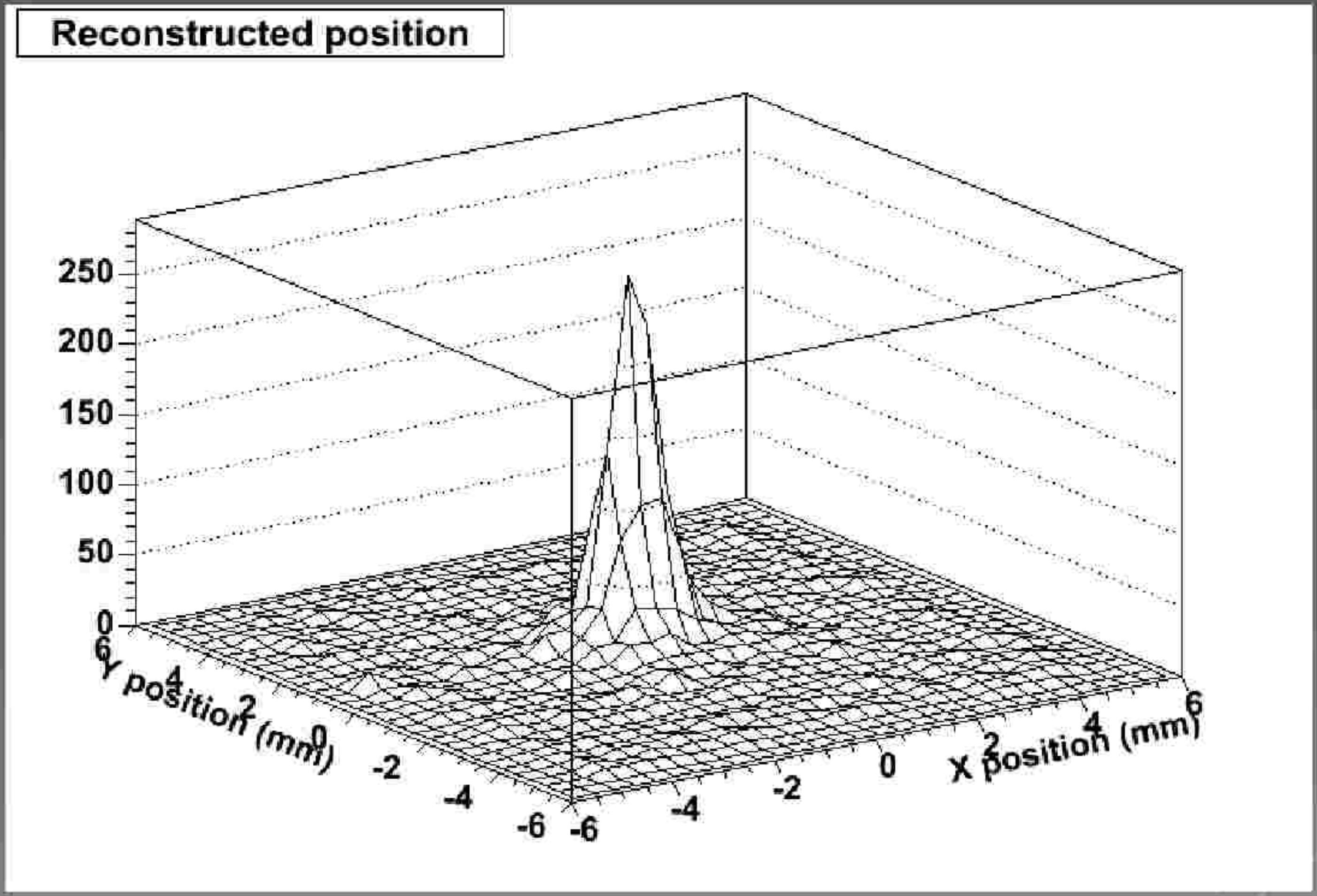
Breakdown voltage over the entire wafer surface

Matrix 8x8 + continuous slab of LYSO (5 mm thick): ^{22}Na energy spectrum



Preliminary data, April 2009, unpublished

Matrix 8x8 + continuous slab of LYSO (5 mm thick): Intrinsic Spatial Resolution



Reconstructed position with center of gravity algorithm. The spatial resolution is about **1 mm FWHM** as obtained with a standard center of gravity algorithm. (Preliminary data, April 2009, unpublished)

CONCLUSIONS

Advantages è MANY! (High Res/DOI/TOFPET/Flexible gemetry/…)

Pitfalls

- **Dependence of Gain on Temperature**
  (80mV/K è <1% G/G/20mV è <4% DG/G/K)
  Needs passive and/or active Temperature control

- Very HIGH granularity è **ASIC is needed**
  e.g. see POSTER Session: Front End electronics [Thursday 17.55]
  “CMOS Analog Front-End Channel for Silicon Photo-Multipliers”
  C.Marzocca (Poli Bari and INFN)

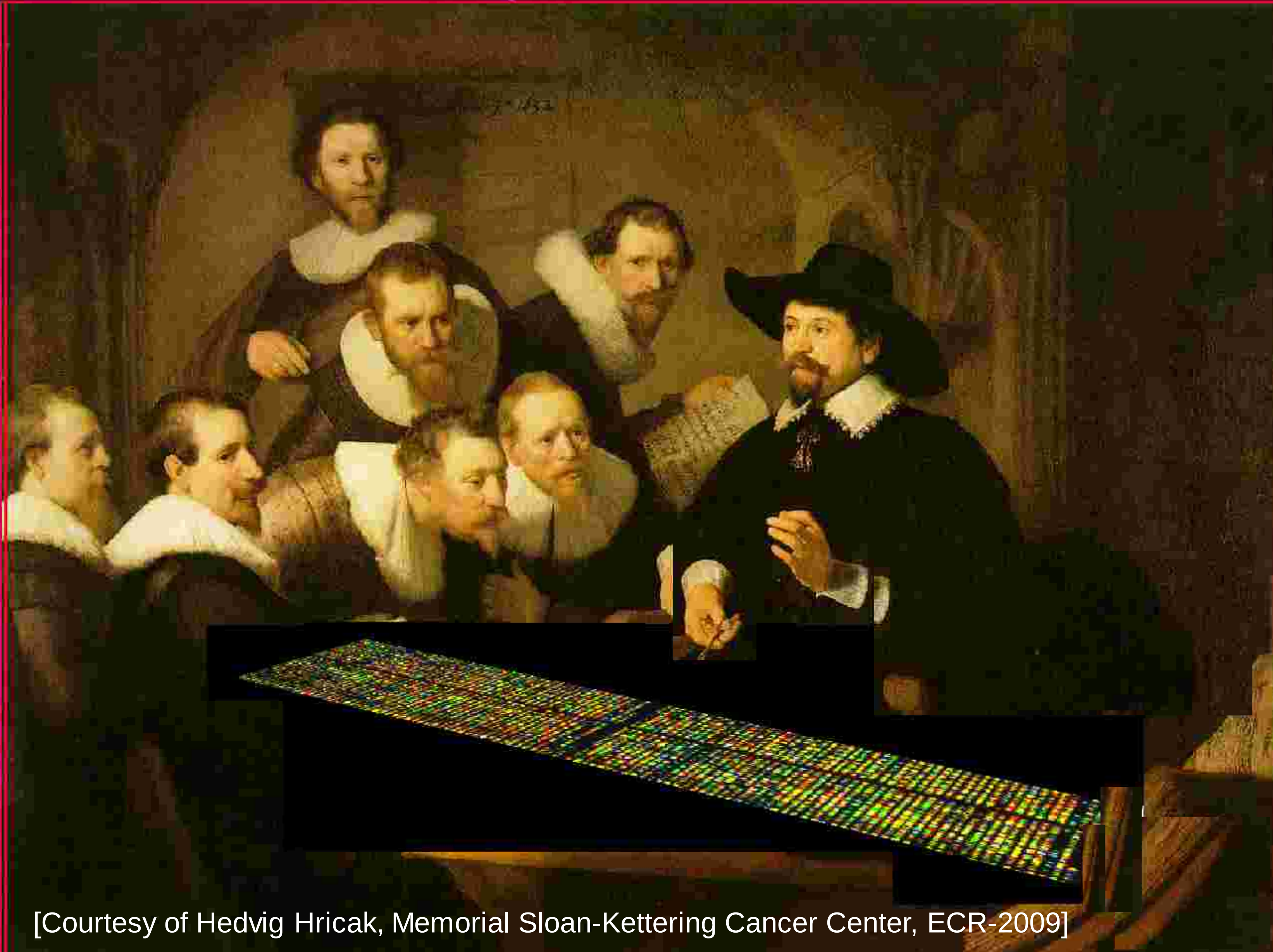
- **10**6 Channels (1x1 mm^2)** for a clinical PET tomograph
  So many working channels for medical physics is a real challenge!

- **Projected Cost for SiPM**:10$ /mm^2
  Compared to PSPMT H8500 è ~2000 euro/25 cm^2 ; ~1$/$mm^2$

All together è MUST be Cost Effective! è Molecular Medicine

ANATOMY LECTURE - 2009 - MOLECULAR MEDICINE

"The Anatomy Lecture of Dr. Nicolaes Tulp" – Rembrandt, 1632



[Courtesy of Hedvig Hricak, Memorial Sloan-Kettering Cancer Center, ECR-2009]

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PRIN 2007- SipmPET (Pisa,Bari,Bologna,Perugia)
Collaboration with Institutions:
ISE srl (Pisa),FBK-isrt (Trento), LAL (Orsay), Univ Cambridge,
Politec Madrid, Univ Valencia, ...

To conclude ….
"A personal VIEW about SIPM by the most faithful PET OMER"

HAPPY END