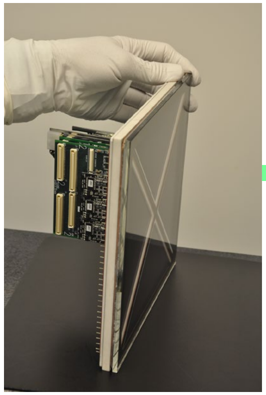
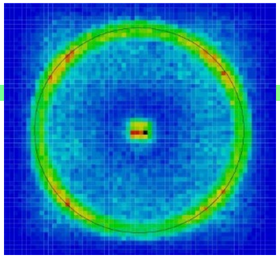
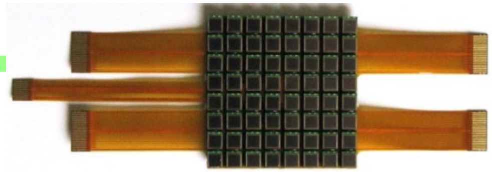
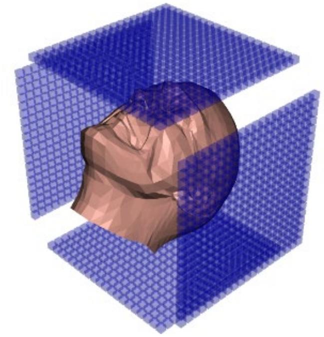
U. Hawaii, seminar, Feb 15, 2024









Instrumentation for advances in PET medical imaging

Peter Križan

University of Ljubljana and J. Stefan Institute

Interplay of detector R&D for particle/nuclear physics and medical imaging

Traditionally excellent collaboration of the two research areas.

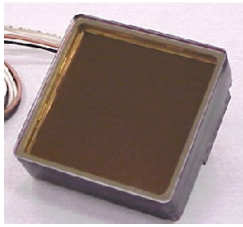
Novel detection techniques required in particle physics →with modifications often applications are possible in medical physics

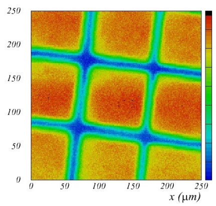
… and sometimes also vice versa…

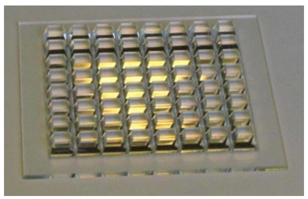
One of the recent examples: SiPMs as scintillation light sensors for

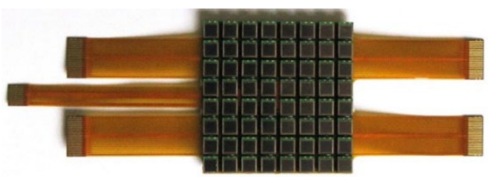
- Electromagnetic calorimeters
- PET scanners

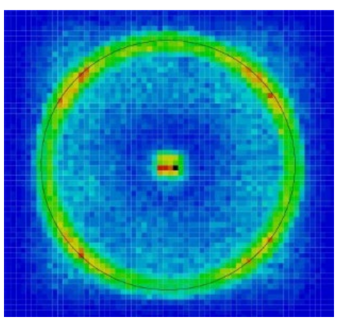
Our original expertise: Cherenkov detectors, single-photon sensors and associated electronics

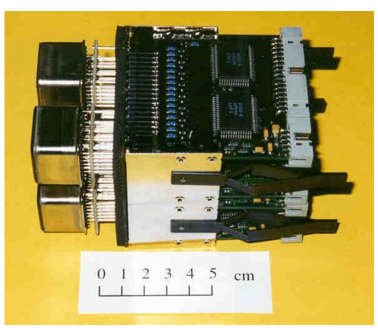


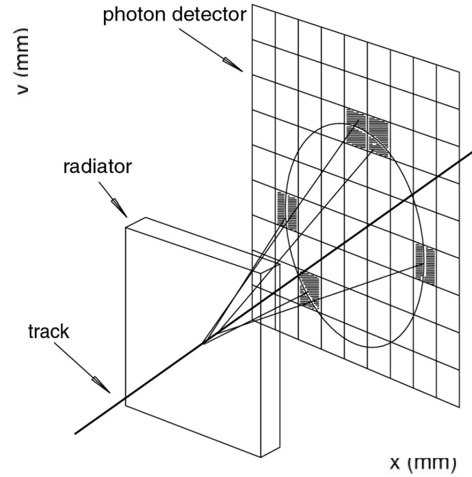


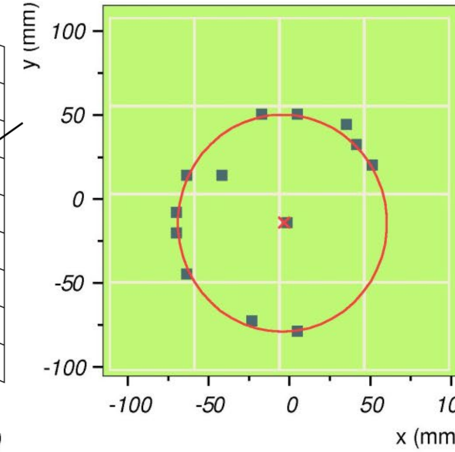


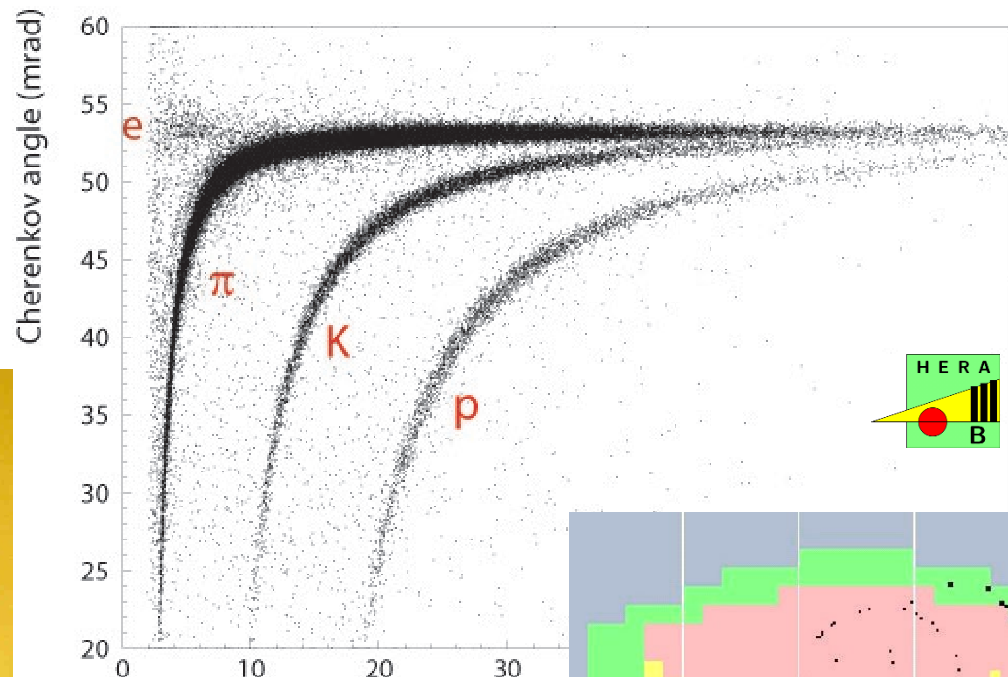


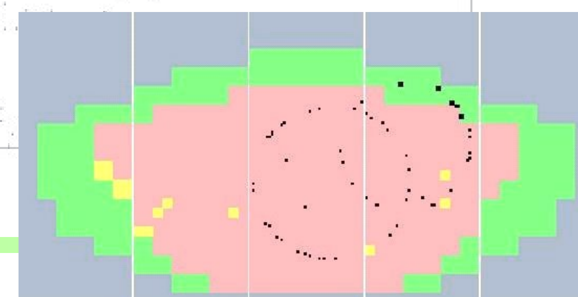




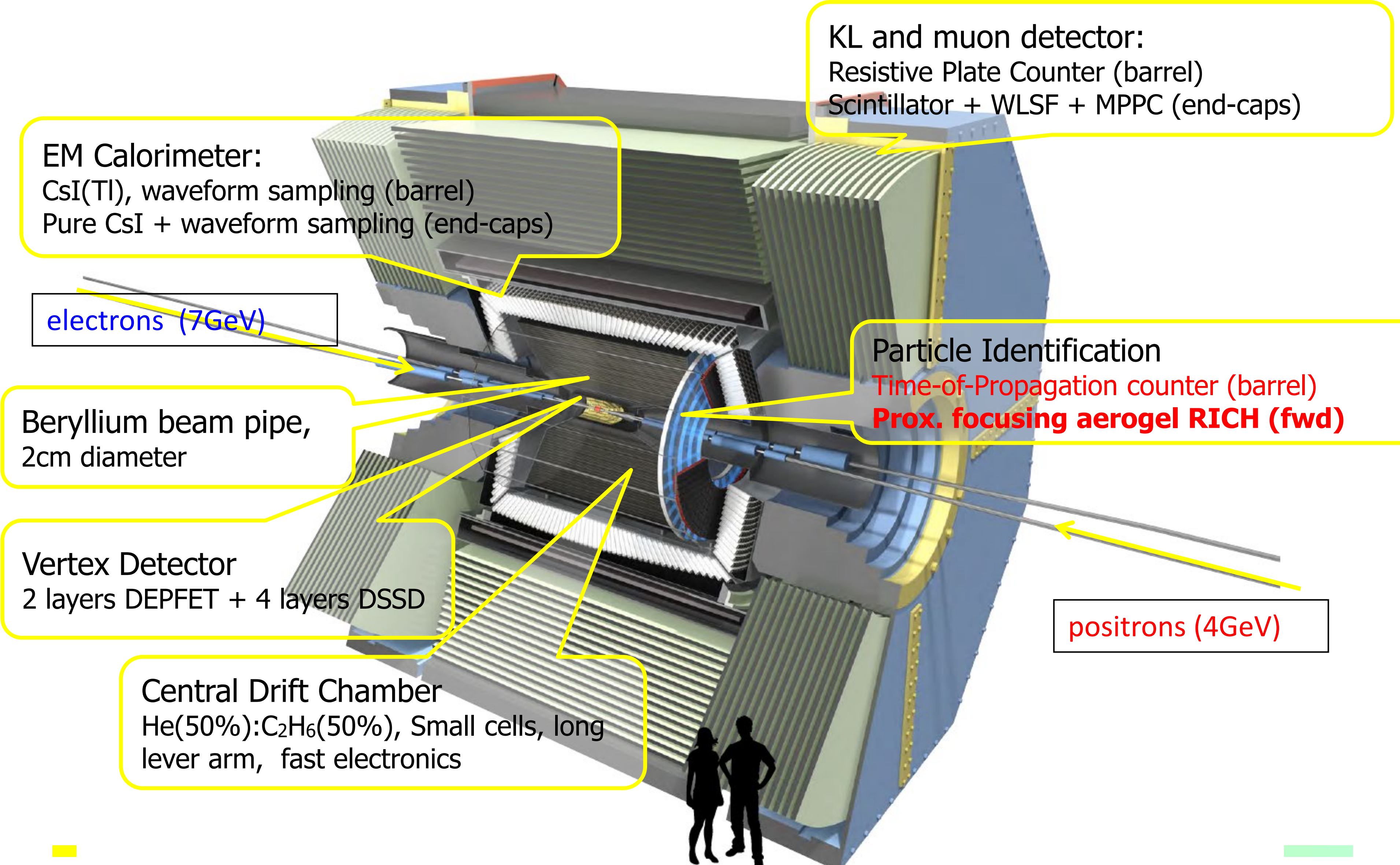








Belle II Detector



Contents

PET – Positron Emission Tomography

Current topics in PET

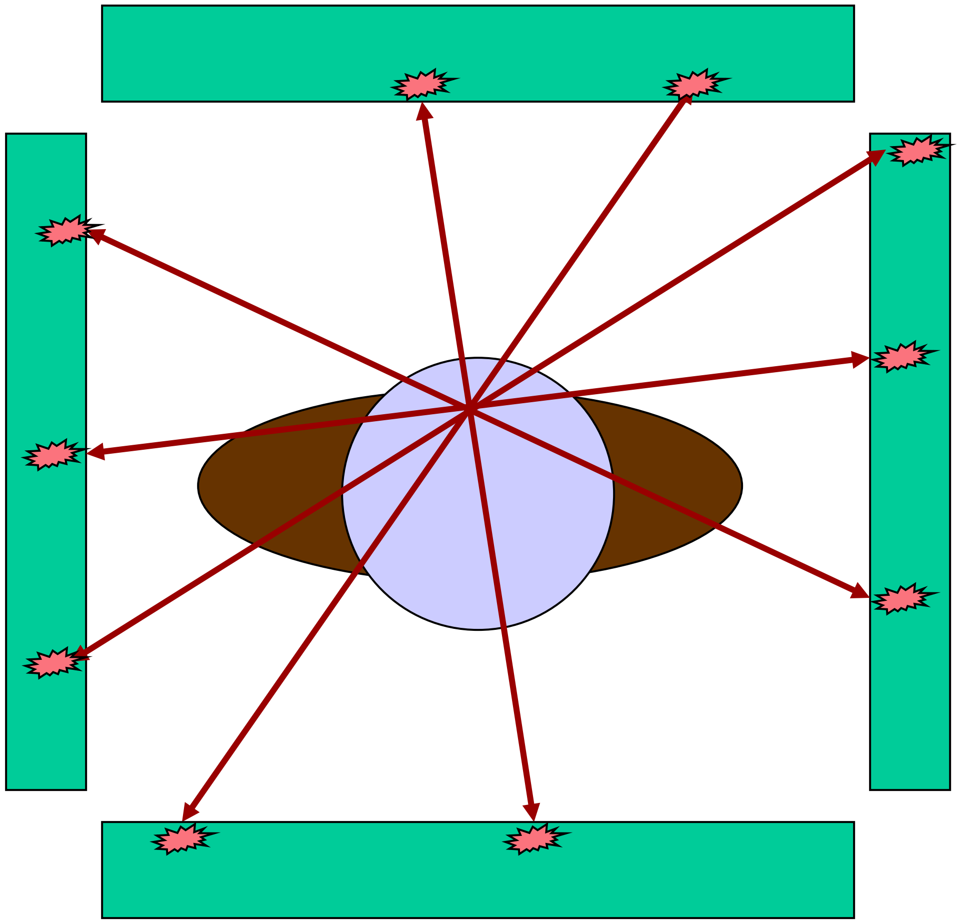
Flexible limited angle PET scanner

Cherenkov radiation based PET scanner

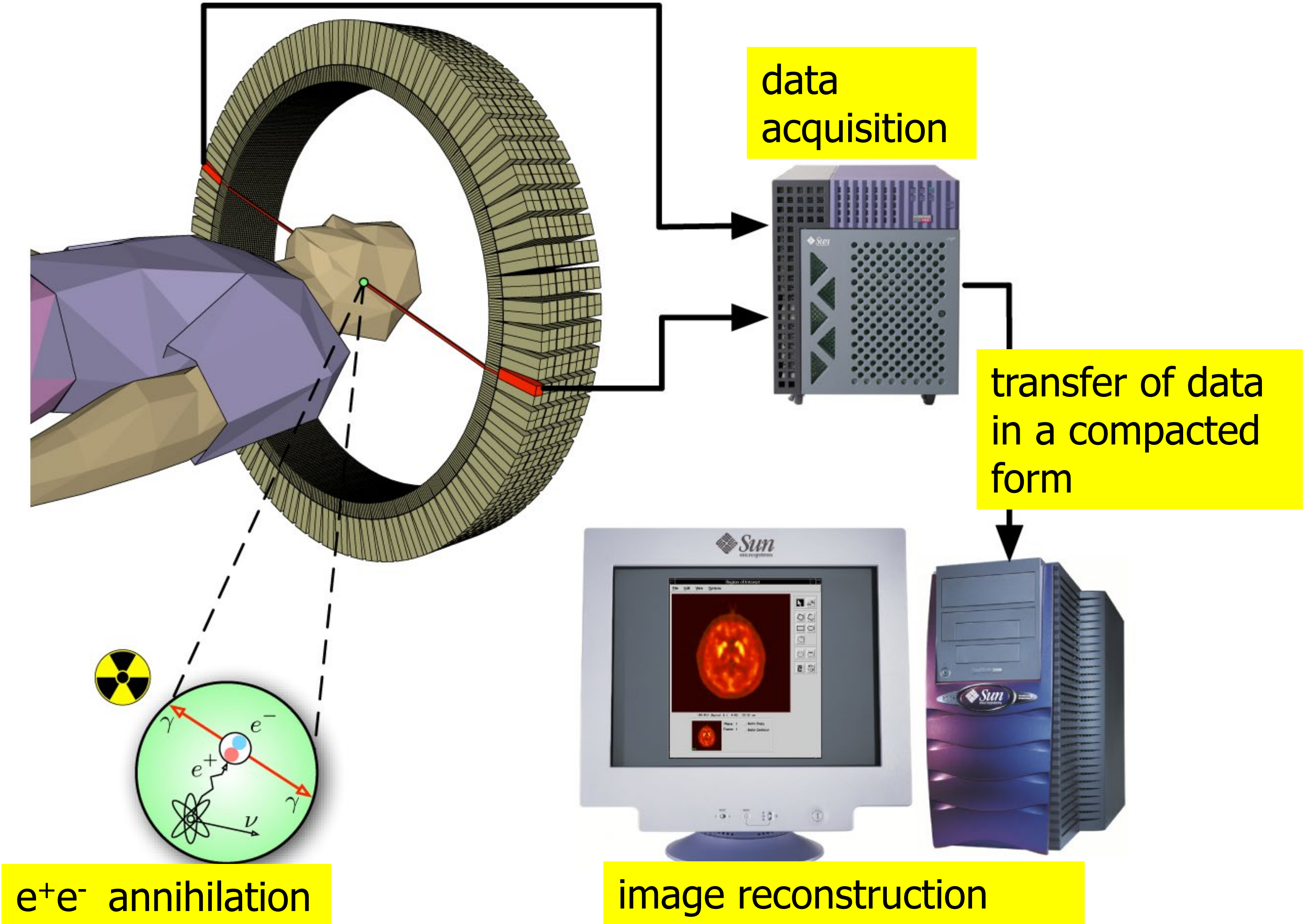
Conclusions and summary

PET: positron emission tomography

In the blood of the patient a substance is administered that contains radioactive isotope – a beta+ emitter (e.g. fluorodeoxyglucose). The places in the body with a higher substance concentration will show a higher activity.

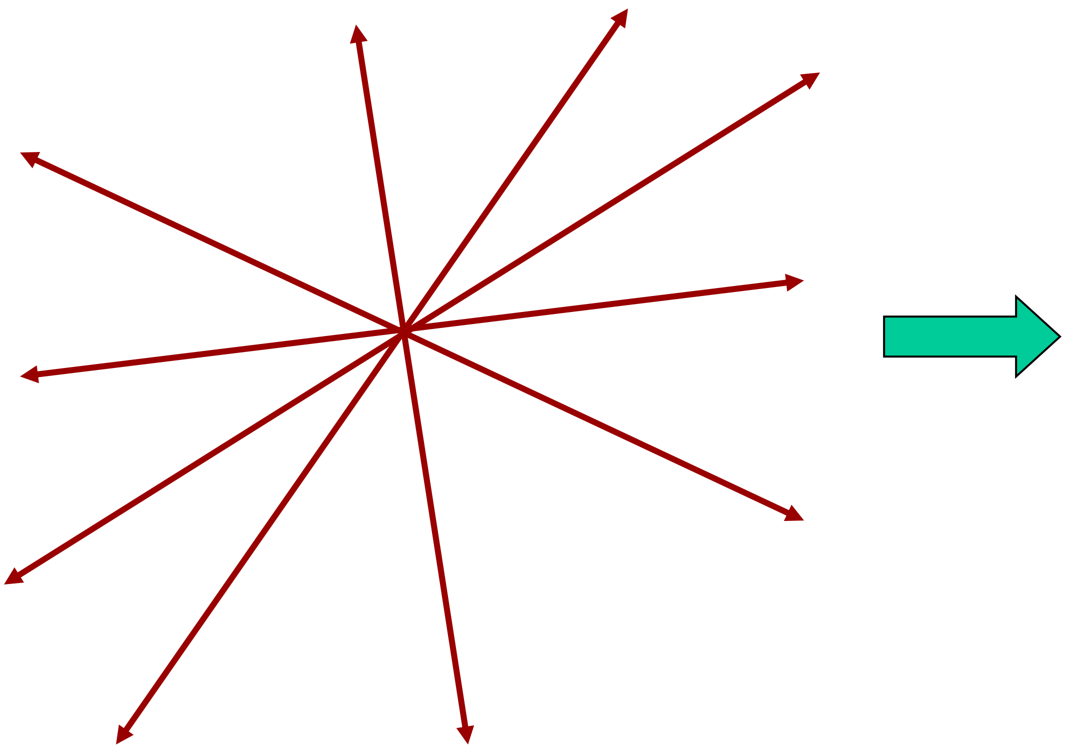


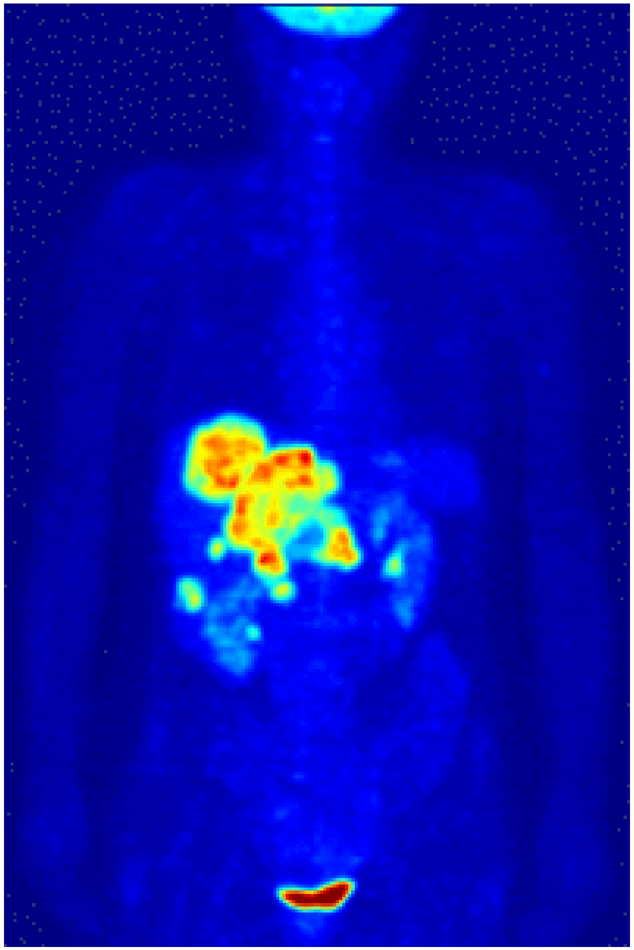
PET: collection and handling of data



PET: image reconstruction

Image reconstruction: from the position and direction of the lines determine the distribution of the radioactive fluorine in the body





PET with a time-of-flight information

Detectors for γ rays can also measure the time of arrival of each of the gamma rays with good enough precision (<1ns)

→ an additional constraint on the point of origin of the two γ rays along the line connecting the two detector hits

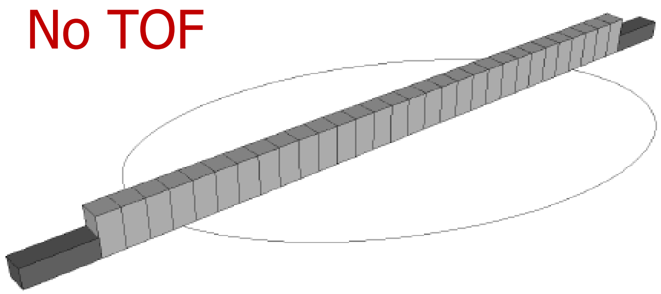
→ time-of-flight (TOF) PET

Good resolution in time-of-flight → limits the number of hit pixels along the line connecting the two detector hits

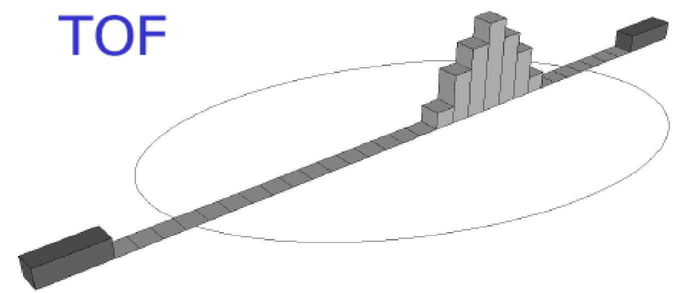
In the reconstruction step, each line contributes to fewer pixels

→ less noise in the reconstructed image

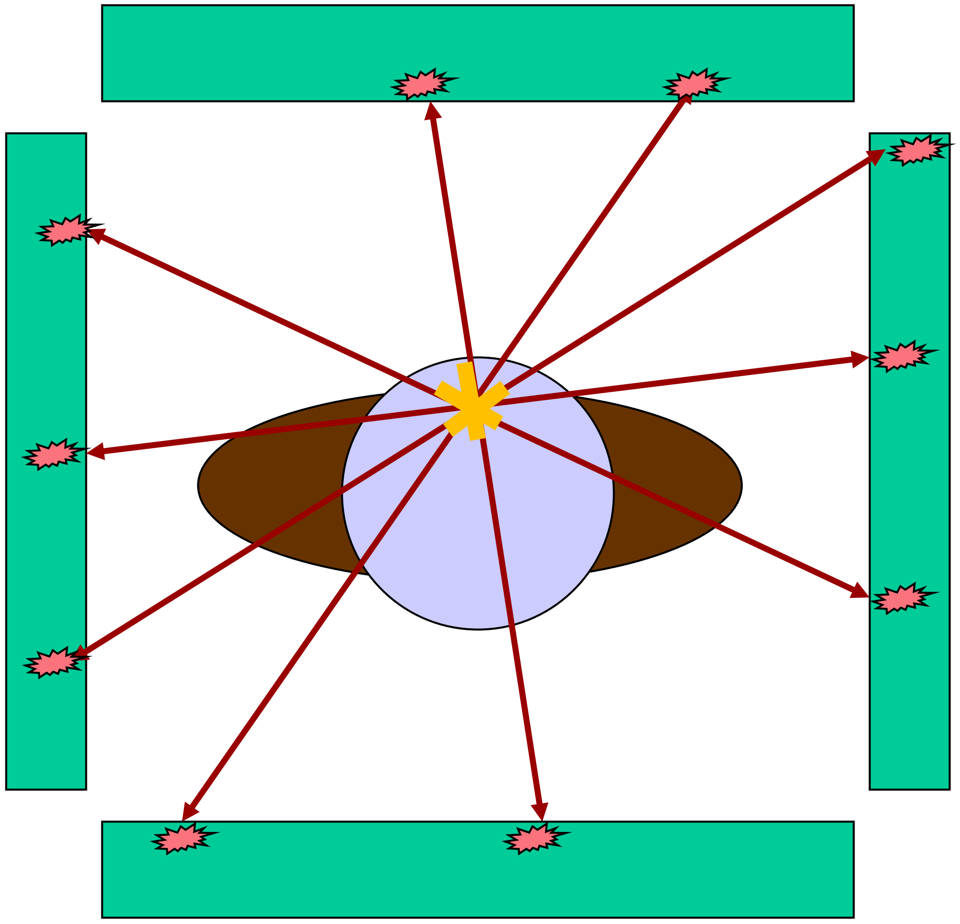
No TOF



TOF

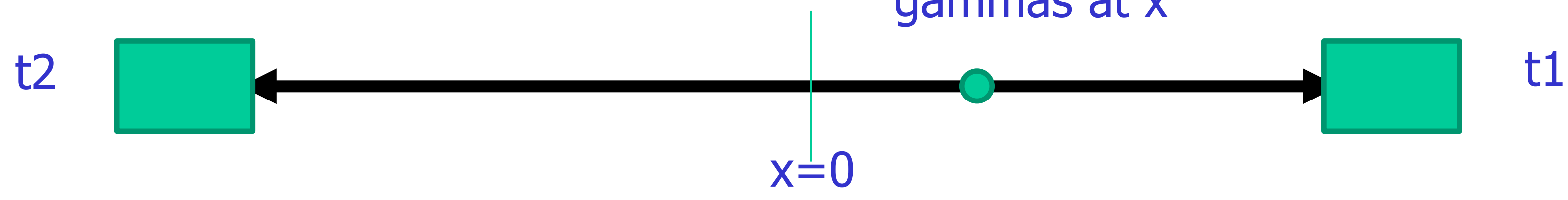


TOF-PET: positron tomography with a time of arrival measurement



TOF-PET: time resolution

What kind of time resolution is needed?



t1 = (L/2 – x)/c

t2 = (L/2 + x)/c

t1 – t2 = 2x/c

source at x, distance between detectors = L

x = (t1 – t2) c/2 → Δx = Δ(t1-t2) c /2

resolution in TOF

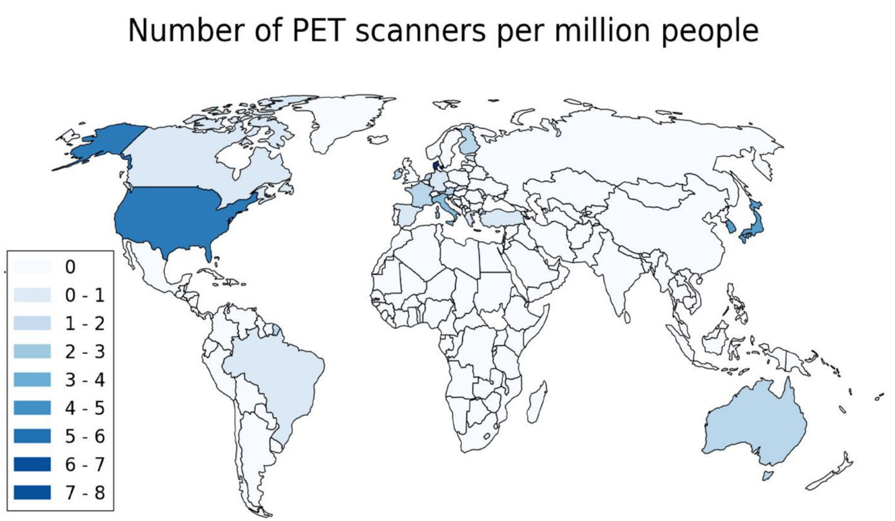
Δ(t1-t2) = 300 ps → Δx = 4.5 cm

Δ(t1-t2) = 66 ps → Δx = 1 cm

Δ(t1-t2) – coincidence timing resolution, CTR

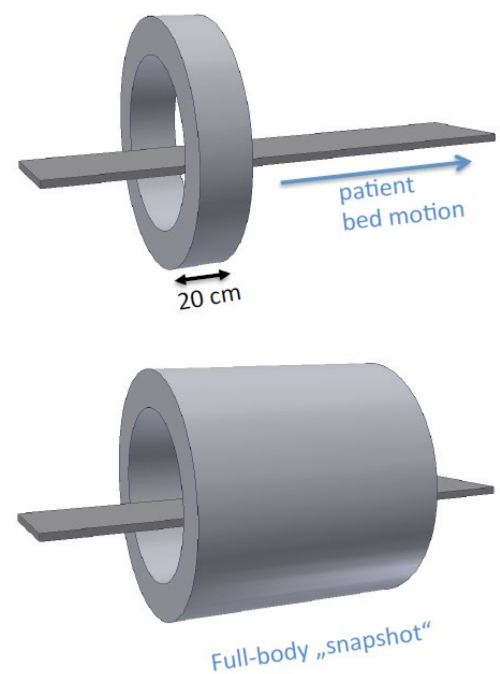
Motivation for Fast TOF PET

- Paradigm shift in medicine from:
  - From the treatment of an obvious disease
  - to early diagnosis / prevention
- This leads to more stringent requirements on PET
  - Sensitivity
  - Specificity
- Targeted Radionuclide Therapy (TRT) & Theranostics
  - introduced an urgent need for more widespread and accurate PET



Current situation

- Standard clinical scanners are sub-optimal:
  - Cost of equipment, limited access, performance.
- Novel long axial PET scanners offer a very attractive solution in terms of
  - increased sensitivity and
  - enabling fast pharmacokinetics/pharmacodynamics.
- They pose significant challenges both
  - Financially
  - Logistically



State-of-the-art in TOF PET

Essential parameter: CTR – coincidence timing recolution

https://www.siemens-healthineers.com/molecular-imaging/pet-ct/biograph-vision

- Clinical scanner:
  - Siemens Biograph Vision PET/CT → **214 ps**

- Laboratory measurement:
  - Gundacker *et al, Phys. Med. Biol.* 65 (2020) 025001 (20pp)

2 x 2 x 3 mm LSO → **58 ps***

2 x 2 x 20 mm LSO → 98 ps*

*measured with single crystals with high power readout electronics that cannot be scaled to large devices

Gamma detectors for PET

Scintillating crystal:
- converts gamma energy into optical photons

Photodetector
- converts optical photons into electrical pulses

Time resolution in TOF PET limited by
- scintillation light emission → 10 ps challenge
  - rise and decay time
- **optical photon travel time spread in the crystal**
- **photodetector response**
- **readout electronics**

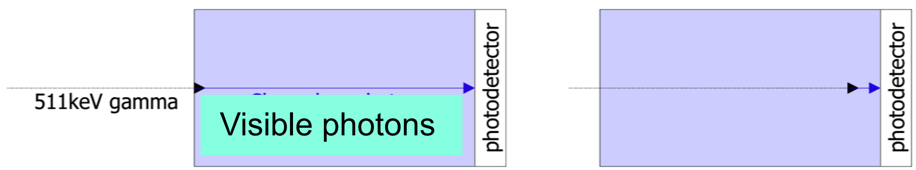
Effective sensitivity $S_{\mathrm{eff},D} \propto \eta_{\mathrm{det}}^2 \, \eta_{\mathrm{geom}} \, \frac{D}{\Delta t}$

- detection efficiency η_{det} of the detector
- η_{geom} the geometrical efficiency (angular coverage)
- D the diameter of the object imaged
- Δt coincidence timing resolution - CTR

Important: Optimize detector CTR to maximize sensitivity

Limitations on timing due to optical travel time

- optical photons, produced in the crystal, need to reach the photodetector
- inside the crystal, optical photons propagate at a lower speed (c/n) than gamma rays (c)
- refractive index, crystal dimensions → **intrinsic travel time spread** due to different gamma interaction depths
- for a 15 mm long crystal this contribution is > 40 ps FWHM:



d = 15 mm, n = 1.8: $t = d \cdot n/c_0 = 90$ ps $t = d/c_0 = 50$ ps $\rightarrow \Delta t = 40$ ps

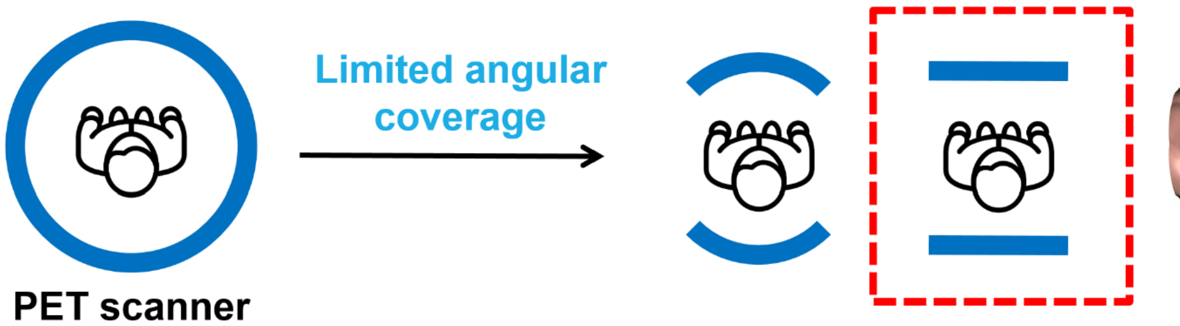
- Can in principle be corrected for by:
  - measuring the depth of interaction (DOI)
  - building the detector with shorter crystals → multi-layer configuration

Can we simplify the TOF PET scanner

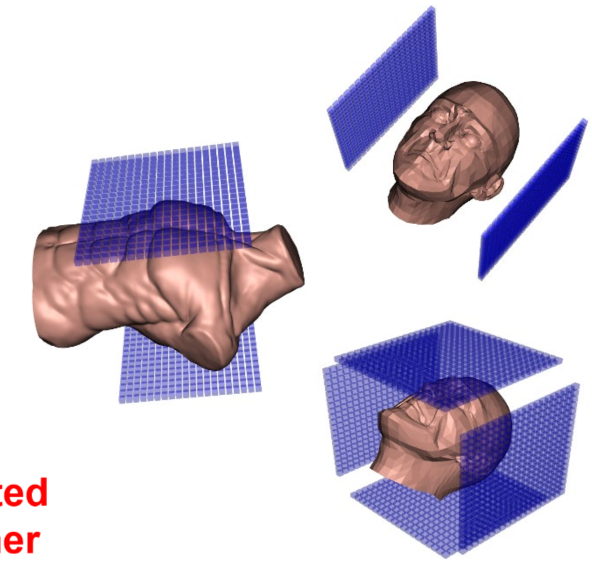
– and make it cheaper and flexible?

Next generation scalable time-of-flight PET

Superb time resolution enables simplifications in the scanner design



Panel-based limited angle PET scanner



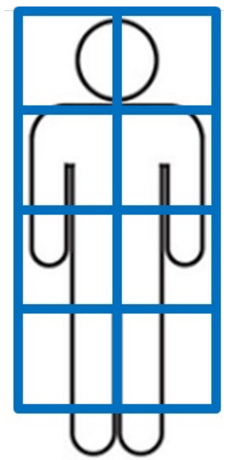
Limited angle PET scanners will generally produce distorted images with artefacts - unless they have good **time-of-flight** information

The angular sampling requirement to obtain distortion-free images decreases
S. Surti, J. S. Karp, Physica Medica 32 (2016) 12–22

G. Razdevšek *et al.*, "Multi-panel limited angle PET system with 50 ps FWHM coincidence time resolution: a simulation study," in *IEEE TRPMS*, doi: 10.1109/TRPMS.2021.3115704.

Potential benefits

- Mobility
  - Portable or bedside PET imaging
- Flexibility
  - Adjustable FOV and sensitivity
- Modularity
  - Combining multiple panels → multi-organ/total-body PET scanner
- Accessibility
  - Reduced manufacturing cost and complexity



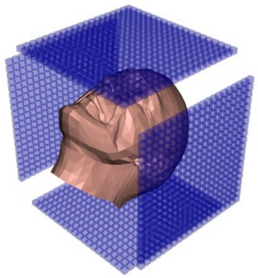
Simulation of a limited angle system

Geant4/GATE → Monte Carlo simulations of digital phantoms and different scanner designs

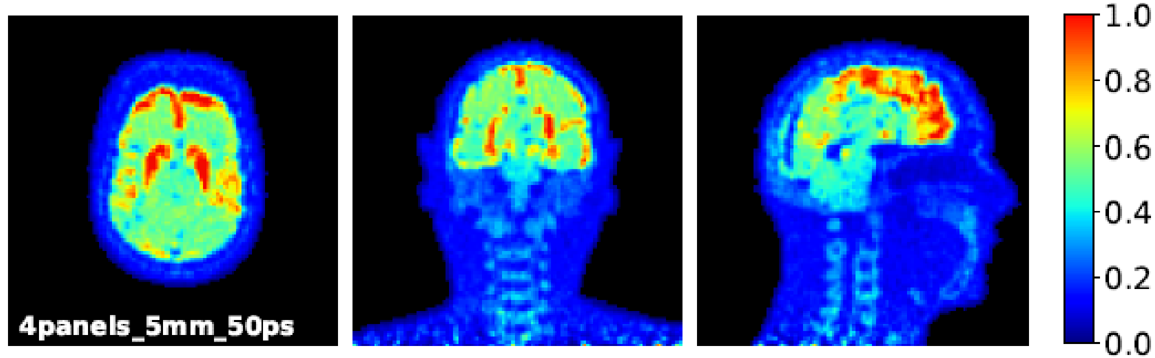
CASToR → image reconstruction with Maximum Likelihood Expectation Maximization (**MLEM**) algorithm

- Investigate the benefits of coincidence time resolutions
- Study the performance **two-panel** and **four-panel** designs

Enabling Open Geometry systems

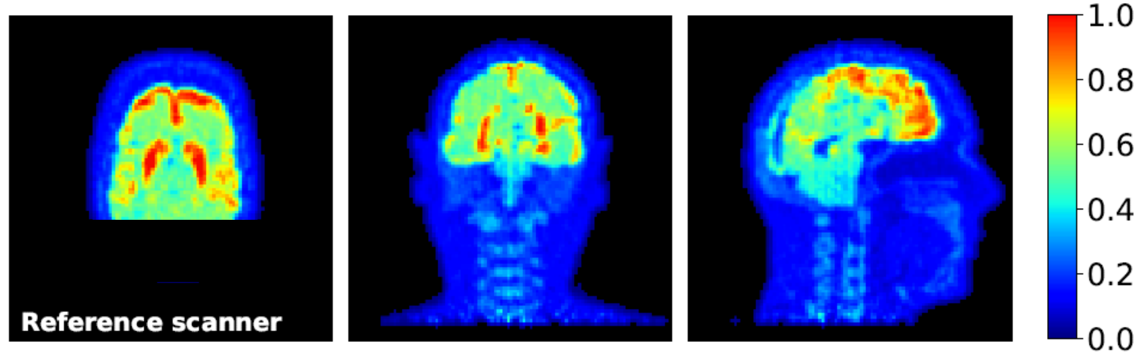


Two pairs of 30x30 cm2
LYSO panels with 50ps CRT

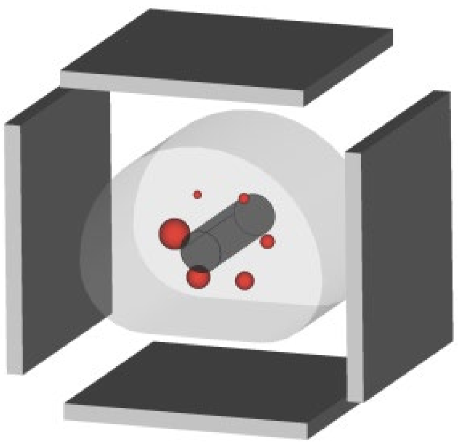


Similar performance as

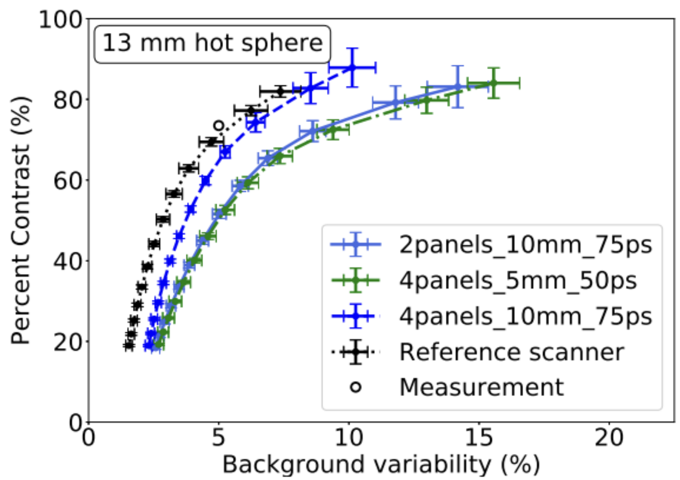
Siemens Biograph Vision



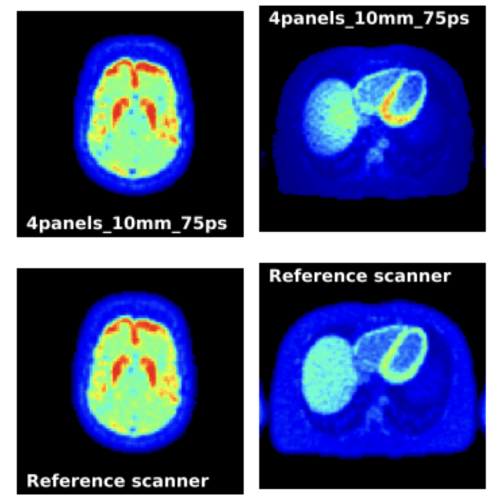
Simulation study of planar configurations



Simulated arrangement of 30x30 cm2 flat panel detectors



Percent contrast versus background variability (~noise level in the image)



Reconstructed images of a torso and head for the flat panel detectors and the reference scanner Siemens BV

G. Razdevšek *et al.*, "Multi-panel limited angle PET system with 50 ps FWHM coincidence time resolution: a simulation study," *IEEE TRPMS*, doi: 10.1109/TRPMS.2021.3115704.

Next generation scalable time-of-flight PET

Address PET challenges of a limited anguar coverage using fast CTR

Joint effort: JSI, FBK, ICCUB, I3M, Oncovision, TU Munich and Yale

- Front-end electronics: develop a low-noise, high-dynamic-range ASIC with a time resolution of 20 ps & on-chip TDC
- Improve SiPM sensor
- Explore 2.5 D integration with the photo-sensor to achieve sub-100 ps CTR

Aim: Improve (SNR) without increasing cost associated with axial coverage by resorting to very sparse angular coverage of the patient and long axial field coverage

Managed to get a 3 MEUR EU grant for 5y to further develop the method and construct a prototype ☺

PET VISION

David Gascon

Institut de Ciències del Cosmos
UNIVERSITAT DE BARCELONA
ICCUB

Chip design

Alberto Gola

FONDAZIONE BRUNO KESSLER
FUTURE BUILT ON KNOWLEDGE

Photo sensors

Jose Benlloch

i3M
Instituto de Instrumentación para Imagen Molecular

Readout Electronics
Data Acquisition

Rok Pestotnik (coordinator)

Design
Integration
Reconstruction

Institut "Jožef Stefan" Ljubljana, Slovenija

Jorge Alamo

ONCO VISION

SME, Key market player:
Mechanics & Software

Georges El Fakhri

Yale

Hospital:
Design & Validation

Wolfgang Weber

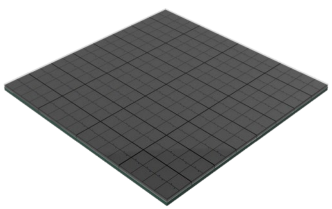
MRI TUM
Klinikum rechts der Isar
Technische Universität München

Hospital:
Validation

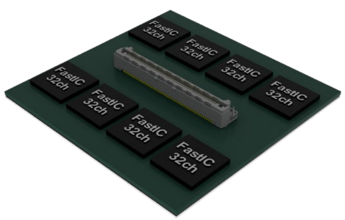
Project Milestones

Jožef Stefan Institute, Ljubljana, Slovenia

Photo Sensor with improved performance

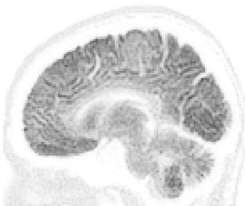


ASIC chip and integration into a digital module



Integration of the prototype

Validation in two hospitals



Exploitation
Certification
Product launch
Bussiness model:
- License to our partner Oncovision (key market player)
or
- Start-up company

Year 2 → Year 3 → Year 4 → Year 5 → Year 6-7

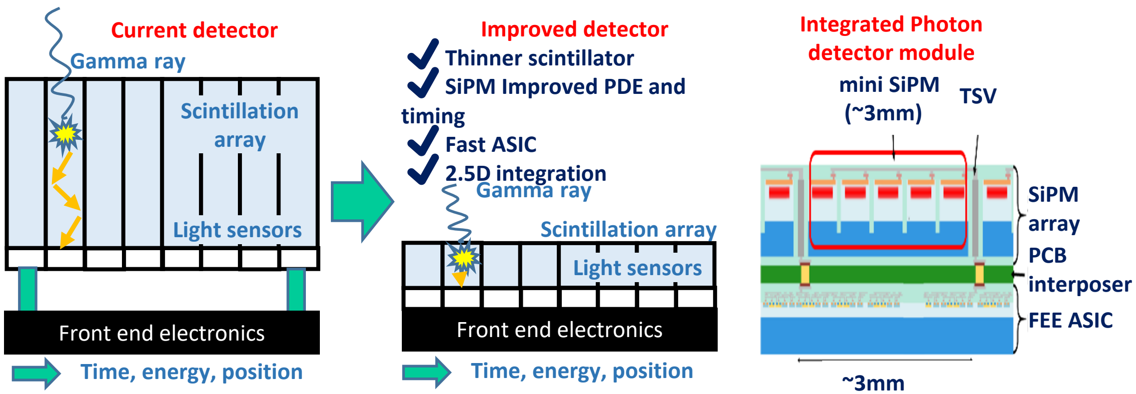
Further funding: EIC Transition, Accelerator, ...

Rok Pestotnik, Jožef Stefan Institute

PET VISION
Funded by the European Union
This project has received funding from the European Union's Horizon Europe research and innovation programme under grant agreement No 101099896

Fast CTR PET module

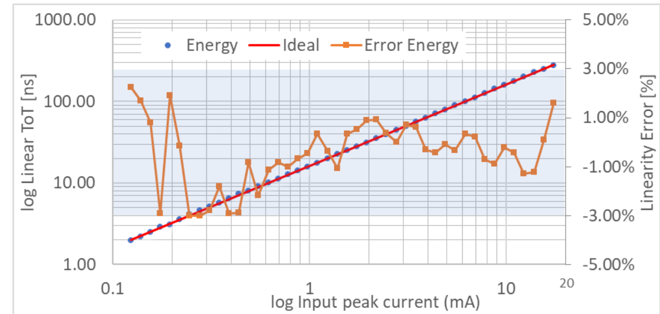
How do we plan to achieve such a good CTR?

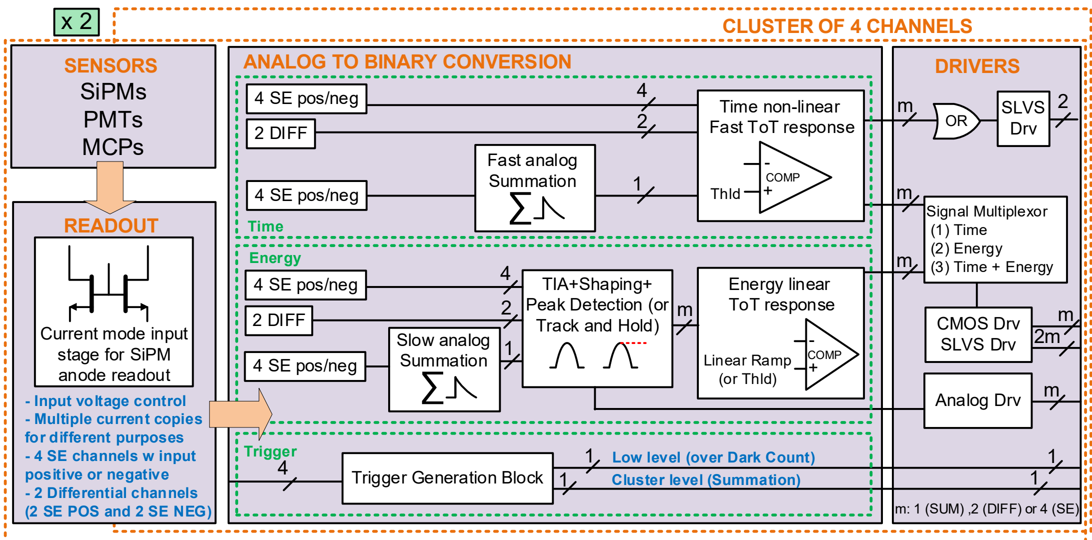


FastIC readout chip

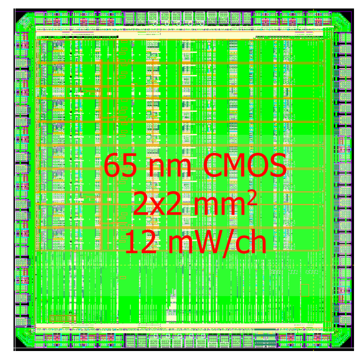
FASTIC: ASIC for fast single photon sensors

- **Collaboration of ICCUB (Univ. Barcelona) and CERN**
- **8 Inputs:** 8 Single Ended (POS/NEG), 4 differential and summation (POS/NEG) in 2 clusters of 4 channels.
- **3 Output modes:** (1) SLVS; (2) CMOS; and (3) Analog.
- Active analog summation of up to 4 SE channels to improve time resolution





- High dynamic range with linear energy response
- Adapted to different detectors: LYSO/LSO, BGO, Cherenkov, Monolithic, etc

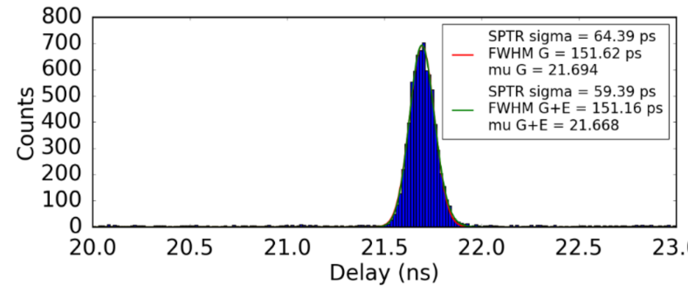


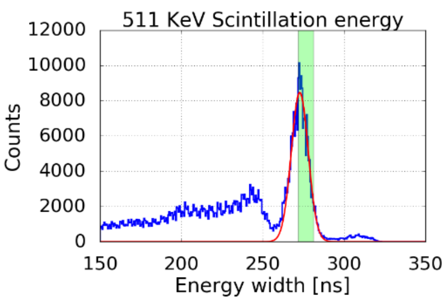
First results with FastIC

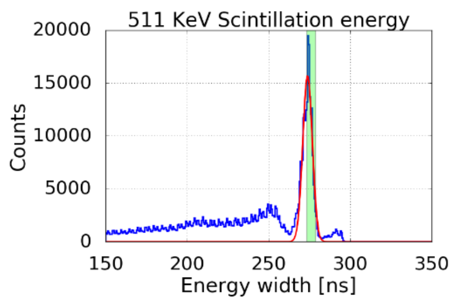
- **Sensor:** FBK-NUVHDLFv2b 3x3 mm², 40 pixel pitch.
- **Crystal:** LSO:Ce Ca 0.2% of 2x2x3 mm³.

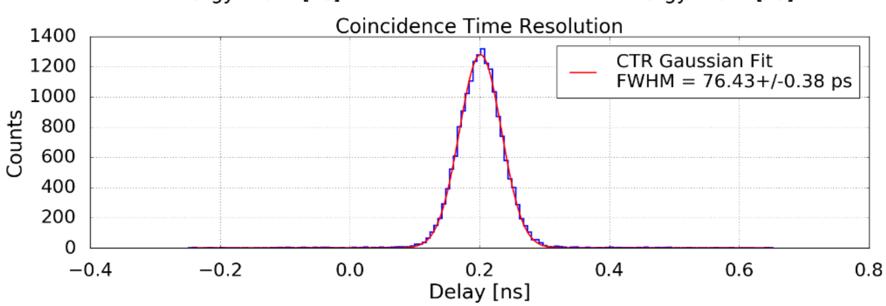
Single photons

- **SPTR with** FBK-NUVHDLFv2b 3x3





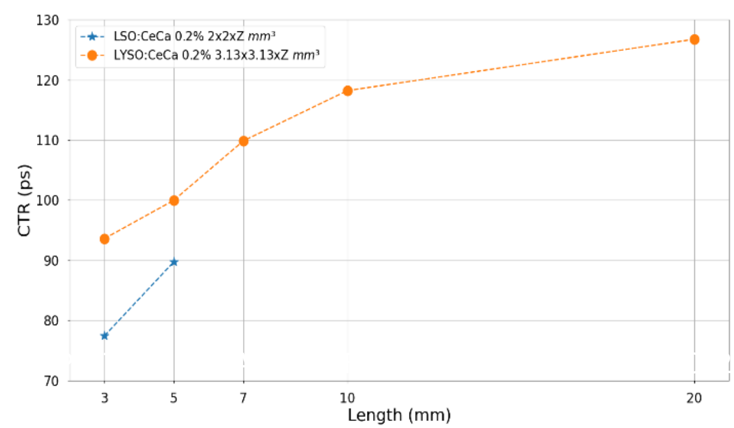




FWHM = 76.43 ps

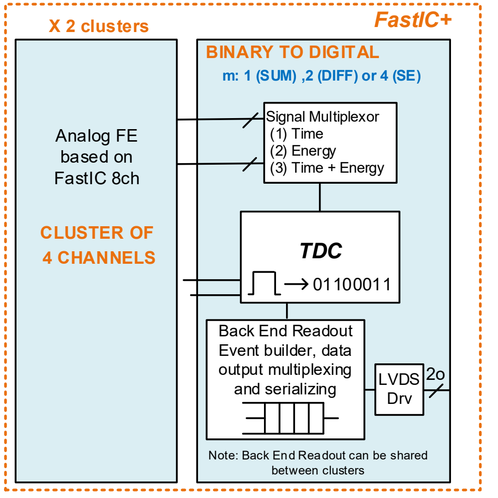
Pairs of annihilation gammas

- **CTR versus crystal length for LYSO and LSO**

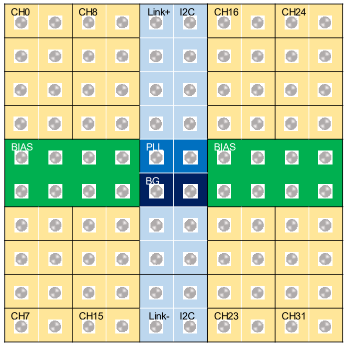


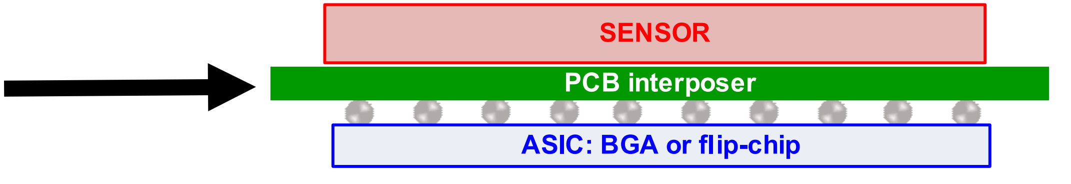
Next generation ASICs

- ICCUB and CERN are working on FastIC+: integration of 25 ps bin TDC integration on FastIC
- On the longer term we plan for a 32 ch. ASIC (FastIC32)
  - Pixelated structure: 2.5D (BGA, flip-chip, etc) or 3D integrated



FastIC 32



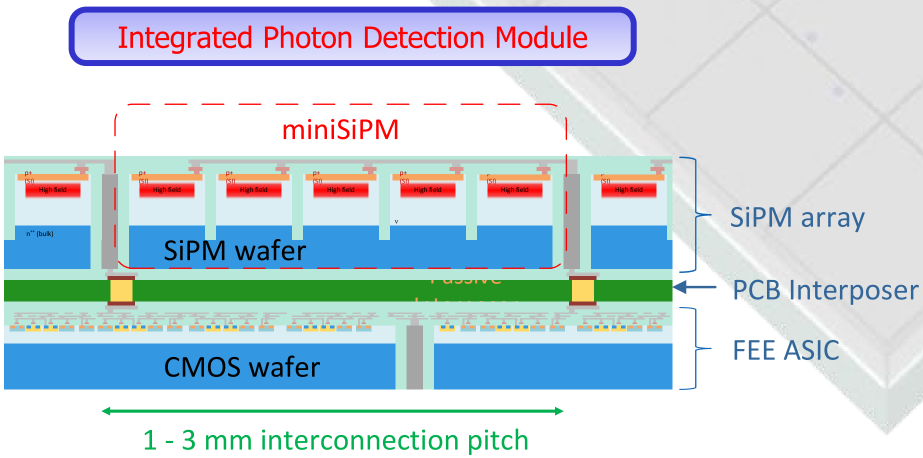


FBK SiPM sensor

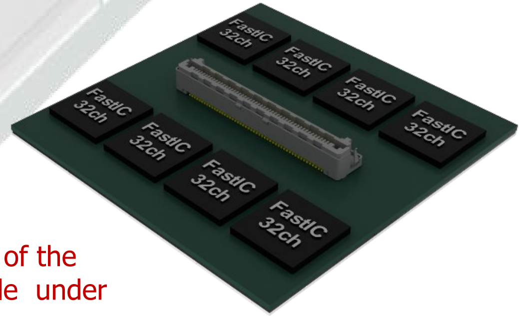
2.5D integrated SiPM tile for improved timing

In the short and medium term - medium density interconnection

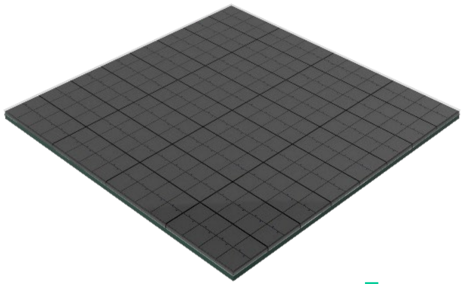
- excellent timing on large photosensitive areas w/o increasing complexity + cost too much.
- SiPMs with TSVs down to 1 mm pitch are connected to the readout ASIC on the opposite side of a passive interposer, in a 2.5D integration scheme.



Hybrid SiPM module being developed for ultimate timing performance in TOF-PET

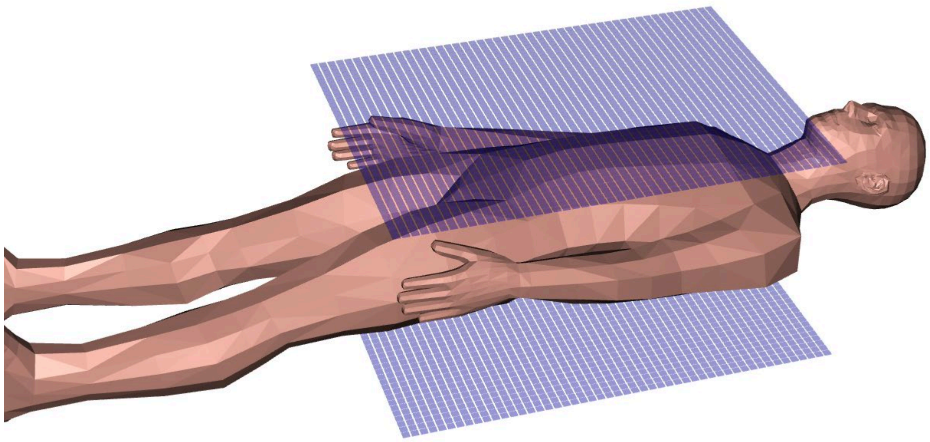


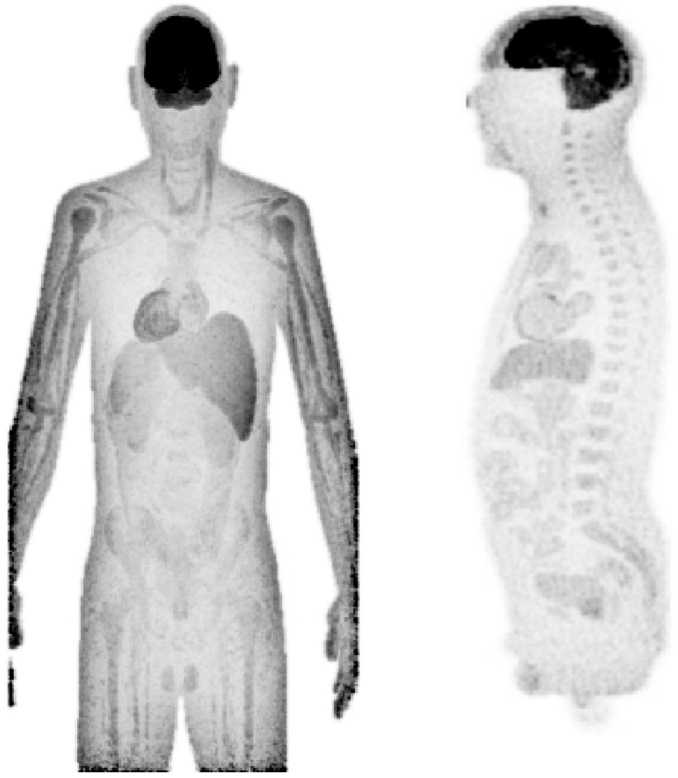
Conceptual drawing of the photon detector module under development



From Limited angle to Total-body

Increased sensitivity by larger panels





Capability of the planar TOF PET imager:
Image of a reconstructed 3 mm slice of an digital phantom acquired by two 120 x 60 cm^2 panel detectors (above and below the patient) assuming 100 ps TOF resolution and 10 mm LYSO scintillator thickness.

Limited angle PET scanner, conclusions

- **Good coincidence time resolution** can:
  - compensate for lower detection efficiency or smaller angular coverage
  - enable us to obtain **good image quality with a simple limited angle PET system** without distortions or artifacts

- We plan to enable open geometry designs and enable a wider spread of PET imaging by reducing different contributions to CTR :
  - Optimize scintillator thickness
  - Improve SiPM – TSV
  - Fast ASIC
  - 2.5D integration
  - If new – faster scintillators emerge, we should be able to make use of them

Use of Cherenkov light in TOF-PET

Use of Cherenkov radiation for TOF-PET

- lead fluoride (PbF_2) as Cherenkov radiator material

Previous work

Limitations of Cherenkov TOF-PET

- single photon detection - **limited scatter suppression**

Image quality with Cherenkov TOF-PET

-whole-body scanner simulations

-crystal readout configurations

-results

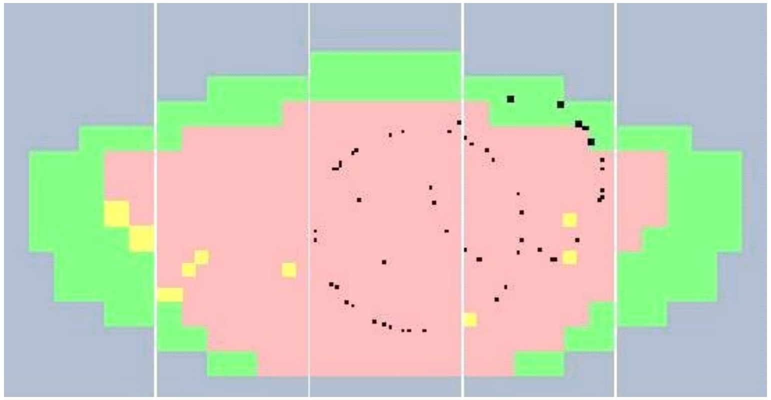
R. Dolenec[a,b], D. Consuegra Rodríguez[a], P. Križan[a,b], M. Orehar[b], R. Pestotnik[a], G. Razdevšek[b], A. Seljak[a] and S. Korpar[a,c]

[a] **J. Stefan Institute**, Ljubljana, Slovenia
[b] Faculty of Mathematics and Physics, **University of Ljubljana**, Ljubljana, Slovenia
[c] Faculty of Chemistry and Chemical Engineering, **University of Maribor**, Slovenia

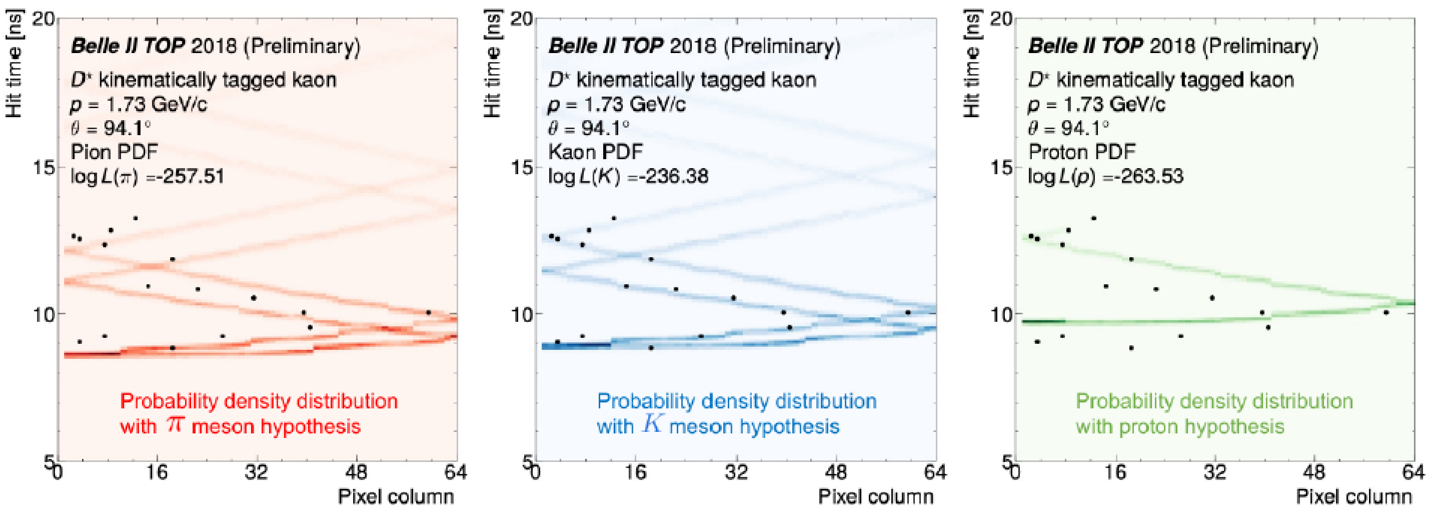
https://photodetectors.ijs.si/

Imaging Cherenkov detectors

Measure the Cherenkov angle (RICH counter)



or a pattern in the coordinate vs time space (iTOP in Belle II)



Use of Cherenkov Light in TOF-PET

γ detectors in traditional PET: scintillator crystal + photodetector

Charged particles (e^- produced by γ interactions) passing trough dielectric material with $v > c_0/n \rightarrow$ **prompt** Cherenkov light

Excellent Cherenkov radiator material: **lead fluoride (PbF_2)**

	BGO	LSO	PbF_2
Density (g/cm^3)	7.1	7.4	**7.77**
μ_{511keV} (cm^{-1})	0.96	0.87	**1.06**
Photofraction for 511 keV	0.41	0.32	**0.46**
Raise time (τ_r)	2.8 ns	70 ps	
Decay time (τ_d)	300 ns	40 ns	
Light yield/511 keV (LY)	3,000	15,000	**10** (‡)

(‡) in the 250-800 nm wavelength interval

PbF_2 properties:

- excellent γ stopping properties
- pure Cherenkov radiator (no scintillations)
- excellent optical transmission (down to 250 nm), high refractive index (n ~ 1.8)
- low price (**1/3 BGO, 1/9 LSO**)

[Mao, IEEE TNS 57:6 (2010) p.3841]

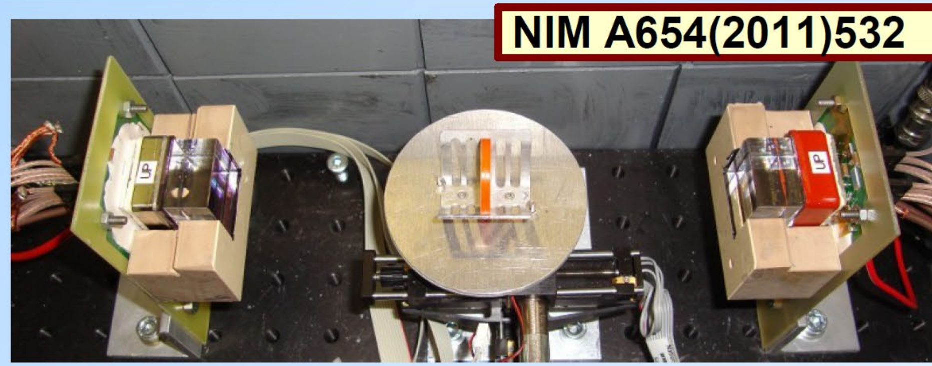
Excellent timing with MCP PMTs*

- Cherenkov radiators:

25x25x(5, 15) mm^3 PbF_2

- MCP-PMT photodetectors:
  - single photon timing ~ 50 ps FWHM
  - active surface 22.5x22.5 mm^2

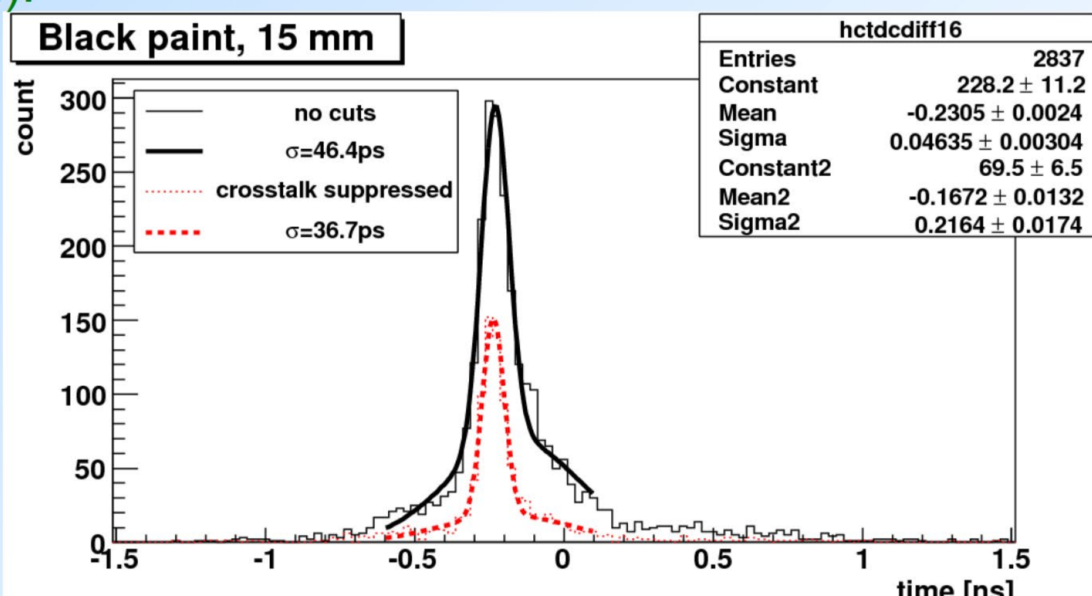
NIM A654(2011)532



black painted, Teflon wraped, bare

- Timing resolution (black painted):
  - ~ 70 ps FWHM, 5mm
  - ~100 ps FWHM 15mm
- Efficiency (Teflon wraped):
  - ~ 6%, single side

(~ 30% for LSO in ideal case)



* The same type of PMTs as employed in the Belle II TOP

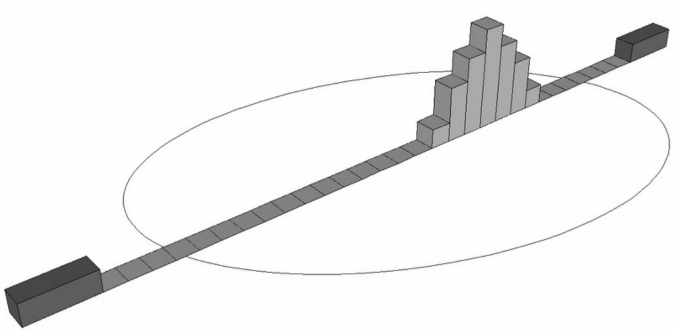
Point source position

Data taken at three ^{22}Na point source positions spaced by 20 mm:

- average time shift 125 ps
- timing resolution ~ 40 ps rms,

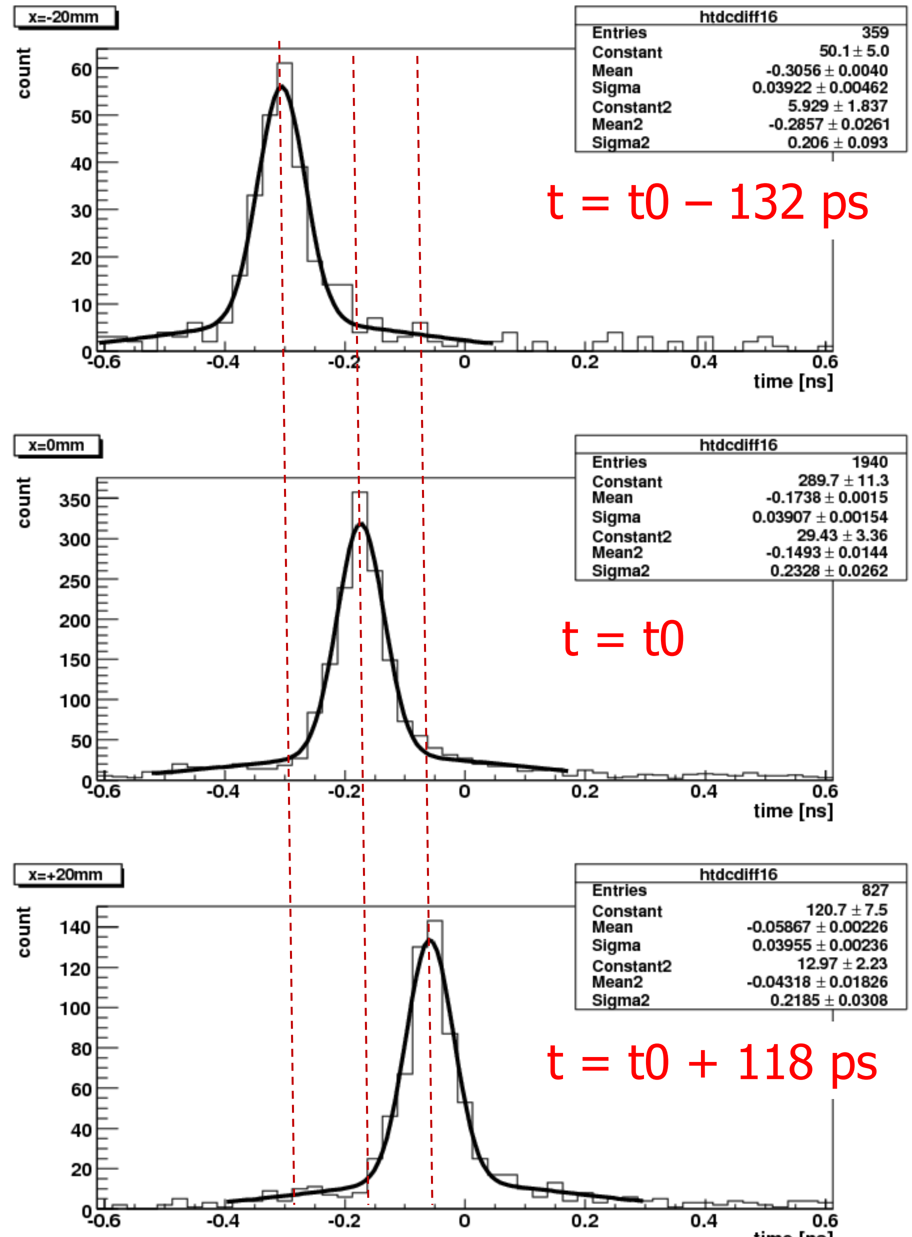
  ~ 95 ps FWHM
- position resolution along line of response ~ 6 mm rms,

  ~ 14 mm FWHM

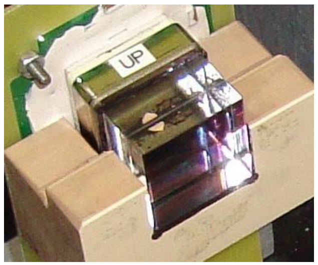


Black painted 15 mm PbF_2 crystals.

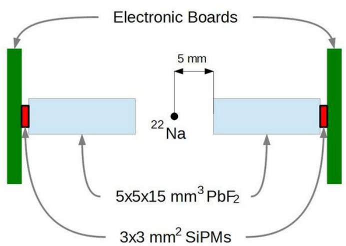
→ NIM A654(2011)532–538

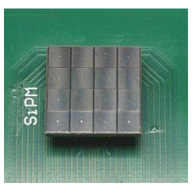


Previous results



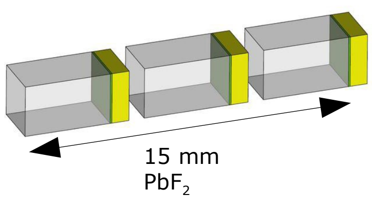
(25 * 25 * **15**) mm³ **PbF_2** (black)
+ (22.5 * 22.5) mm² **MCP-PMT**





4x4 array:
(3 * 3 * **15**) mm³ **PbF_2** (reflector)
+ (3 * 3) mm² **SiPM**

Multi-layer: 3 x
[(3 * 3 * **5**) mm³ **PbF_2** (black)
+ (3 * 3) mm² **SiPM**]



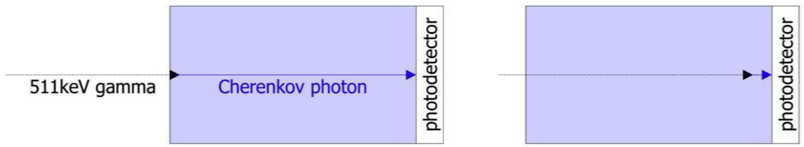
Result	Reference
Cherenkov TOF PET TOF: **95 ps** FWHM	Korpar, NIM A 654 (2011) 532
With SiPMs TOF: **306 ps** FWHM	Dolenec, IEEE TNS 63:5 (2016) 2478
Cherenkov PET module Single side efficiency: **35 %**	Dolenec, NIM A 952 (2020) 162327
Multi-layer detector (simulation) TOF: **22 ps** FWHM before photodetector timing	Consuegra, Phys.Med.Biol. 65(5) (2020) 055013

Limitations of Cherenkov TOF-PET

- Only 10-20 photons created → **only a few detected**
  - efficient photodetector and light collection needed

SiPM

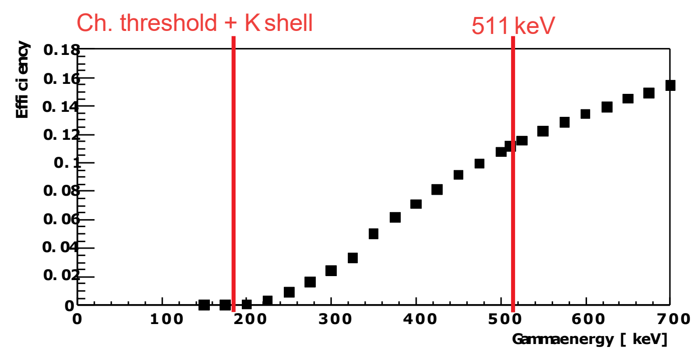
- **Optical photon travel time spread** in the crystal
  - remaining limitation to TOF resolution



d = 15 mm, n = 1.8: $t = d \cdot n / c_0 = 90$ ps $t = d/c_0 = 50$ ps → $\Delta t = 40$ ps

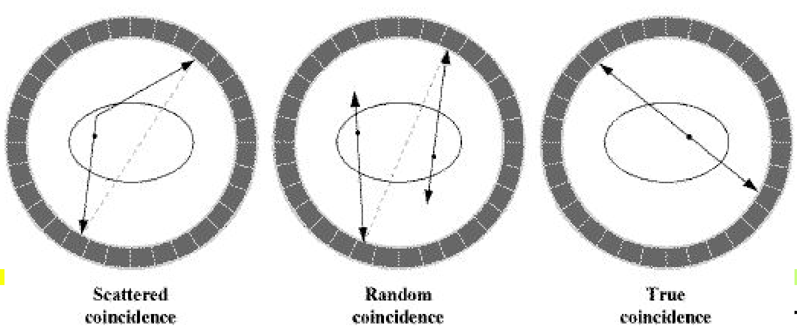
multi-layer

- Limited suppression of **scattered events**:
  - only a few Cherenkov photons detected → no energy information
  - detection efficiency drops at low gamma energies → intrinsic suppression



Simulation, PbF_2 with MCP-PMT photodetector

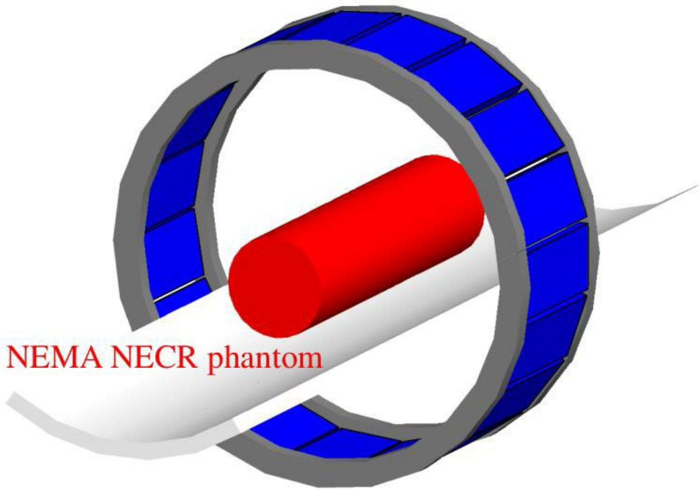
Effect of remaining scatter on image quality?

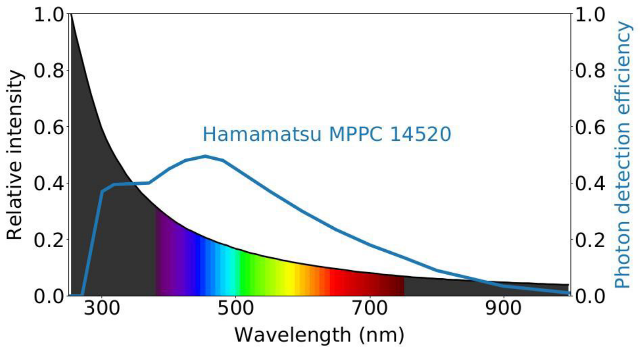


Essential question → MC simulation to evaluate the effect →

Whole-body scanner simulations

- Simulation: GATE v8.1
- Geometry:
  - Based on **Siemens Biograph Vision PET/CT**
    - ring: 19 modules (Axial FOV: 26.3 cm)
    - module: 2 x 8 block detectors
    - block detector: 4 x 2 mini-blocks
    - mini-block: 5 x 5 crystal array
    - **crystal: 3.2 x 3.2 x 20 mm^3**
- Optical simulations (Cherenkov):
  - Surfaces: Geant4 UNIFIED model
    - **reflector** (diffuse, R=95%, n=1.0)
    - **black** (R=0%, n=1.5)
  - Photodetector: **Hamamatsu S14520 SiPM**
    - Single Photon Time Resolution (SPTR): **70 ps** FWHM
    - SiPM dark counts not modeled
- Reconstruction: CASToR v3.1.1
  - Custom double Gaussian TOF kernel [CASToR workshop]
  - OSEM-8it:5sub, 1.6 mm voxel, 5 mm filter



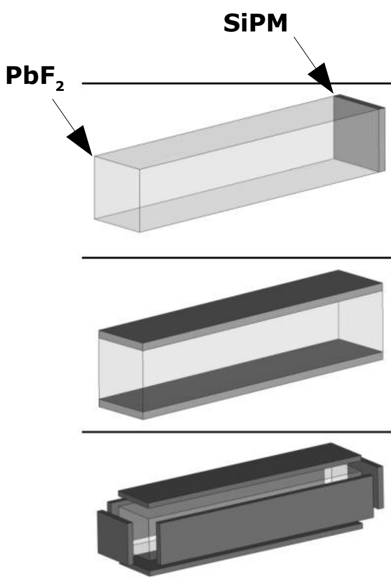


Crystal readout configurations

Reference scanner

- LSO scintillator
- Energy window: 435-585 kev
- Energy resolution: 10%
- CTR: 214 ps

- Cherenkov photon generation, propagation simulated
- Timing defined by first optical photon detected



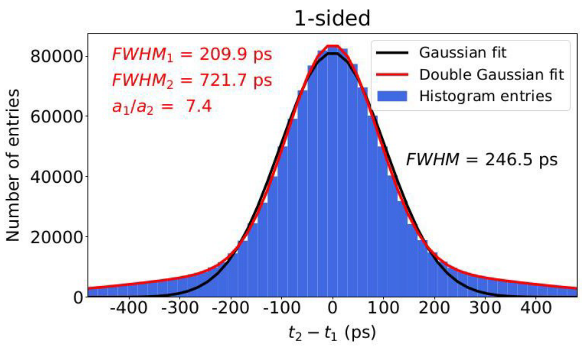
Cherenkov detector	Surface treatment	ε^2 (%)	CTR-FWHM (ps)		FOM	
			0 ps SPTR	70 ps SPTR	0 ps SPTR	70 ps SPTR
1-sided-back	Black	8.6	100.7	145.5	0.85	0.59
	Reflector	35.3	135.7	184.8	2.60	1.91
2-sided-top-bottom	Black	26.2	47.0	111.1	5.57	2.36
	Reflector	40.5	48.9	117.8	8.28	3.44
6-sided	/	44.4	54.1	115.4	8.21	3.85

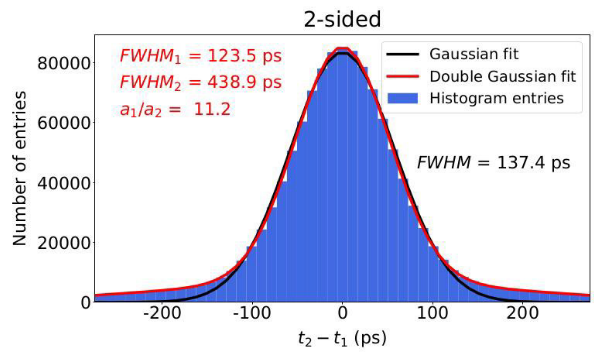
Coincidence detection efficiency: ϵ^2

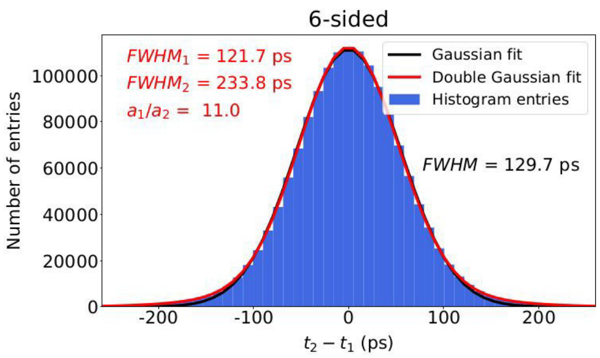
Figure-of-merit: $FOM = \frac{\epsilon^2}{CTR}$

SPTR = single photon time resolution

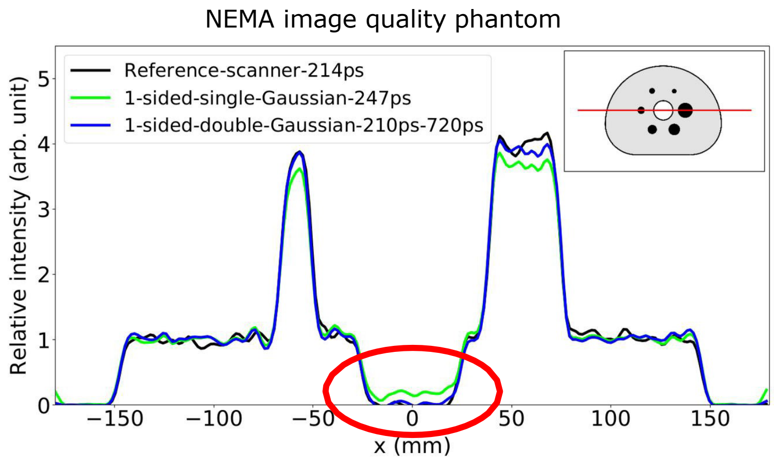
Results: CTR distributions







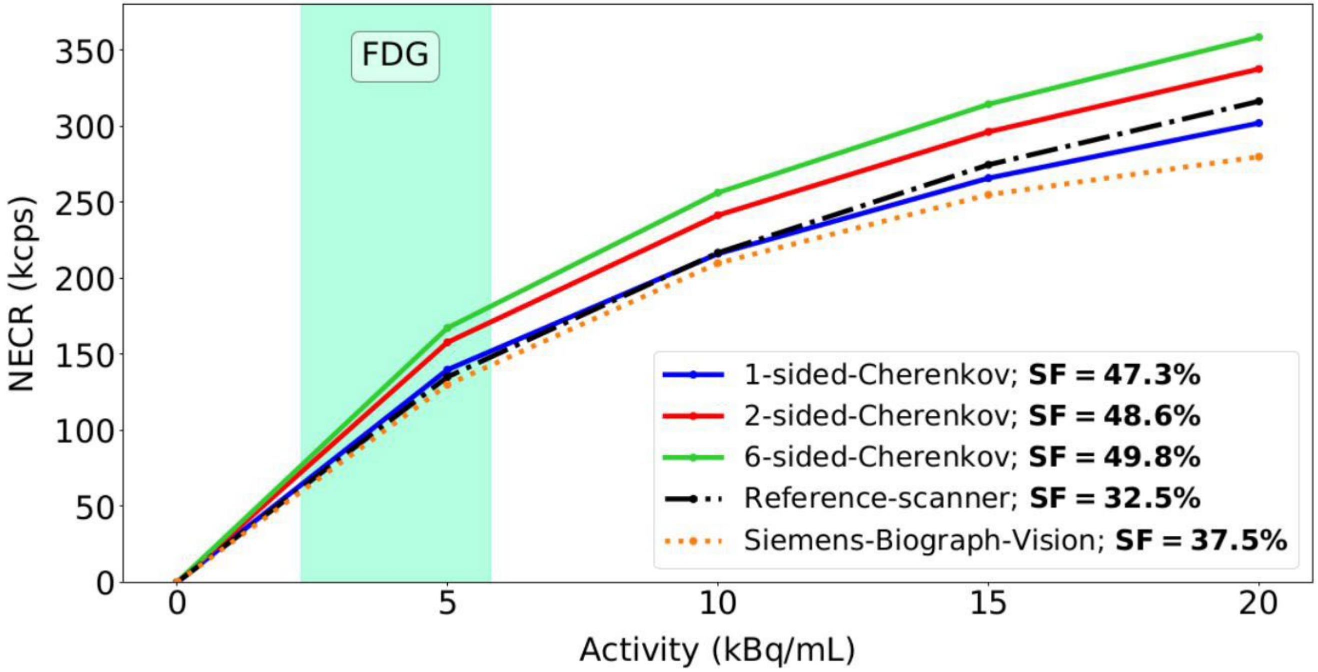
TOF kernel:
single Gaussian
double Gaussian



Results: NECR

- Noise Equivalent Count Rate: $NECR = \frac{true^2}{true + random + scatter}$
  - not influenced by TOF
- Scatter Fraction: $SF = \frac{scatter}{true + scatter}$

NEMA NU 2-2018

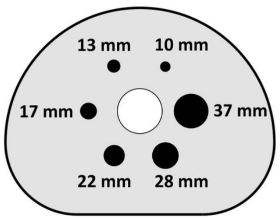


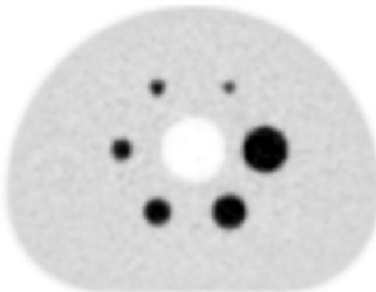
[Sluis, J. Nucl. Med. 60:7 (2019) 1031]

•The "Noise Equivalent Count" is the number of counts from a Poisson distribution (standard deviation estimated by SQRT{N}) that will yield the same noise level as in the data at hand.

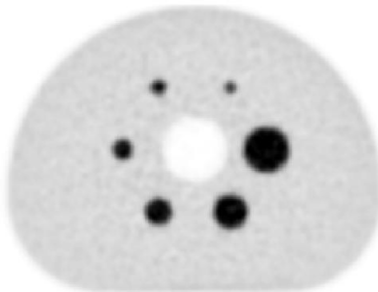
Results: Image Quality

- NEMA image quality phantom



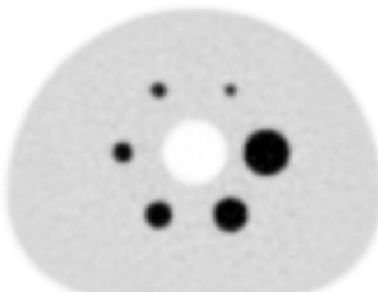


Reference-scanner-214ps

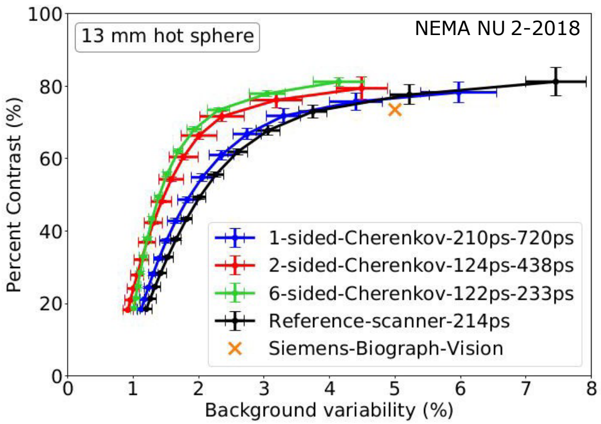


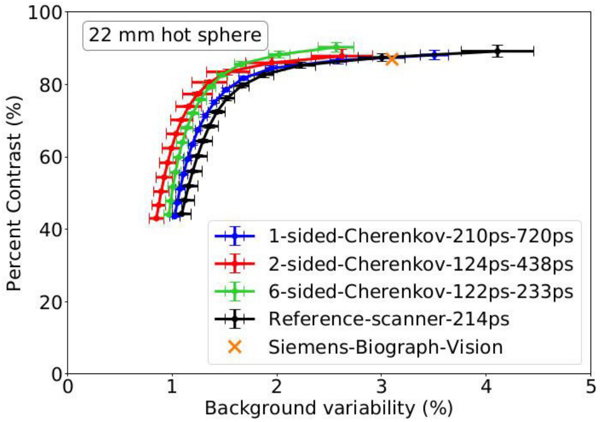
1-sided-210ps-720ps

2-sided-124ps-438ps



6-sided-122ps-233ps

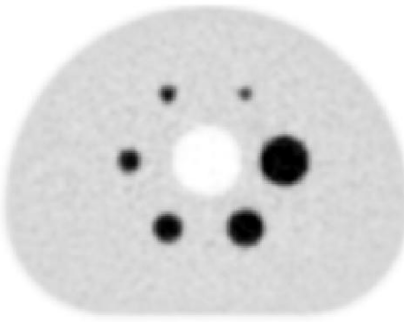




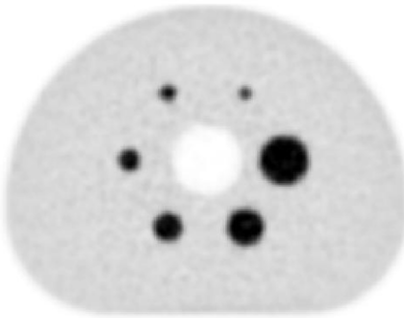
Cherenkov based scanners, conclusion

- Using (exclusively) Cherenkov light in TOF-PET has potential to
  - improve TOF resolution
  - reduce scanner cost (total-body)
- Experiments have demonstrated
  - CTR as low as 30 ps [Ota, Phys. Med. Biol. 64 (2019) 07LT01]
  - detection efficiency (module) of 35% [Dolenec, NIM A 952 (2020) 162327]
- No energy information available → effect on image quality?
- Cherenkov TOF-PET scanner simulations
  - better sensitivity and CTR compensate higher scatter
  - **image quality comparable to state-of-the-art**
- Advanced detector geometries (2-sided top-bottom, multi-layer)
  - even better image quality

[G. Razdevšek et al, IEEE TRPMS (2022)
DOI: 10.1109/TRPMS.2022.3202138]



Reference-scanner-214ps

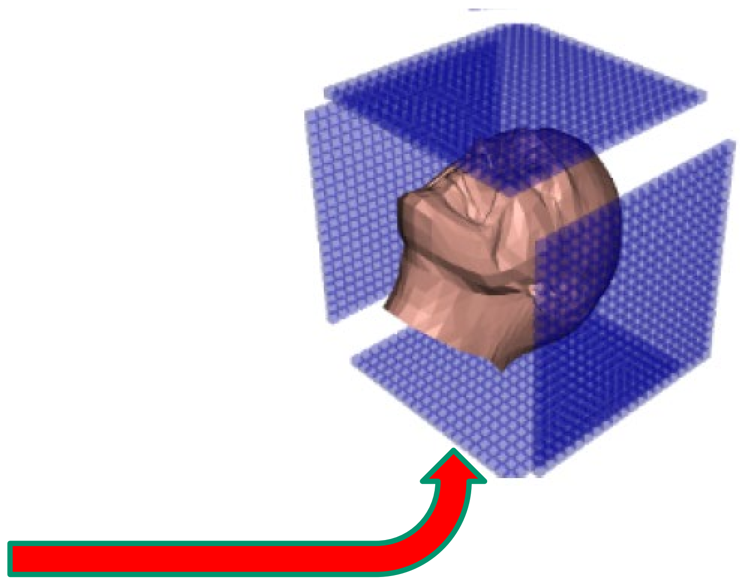


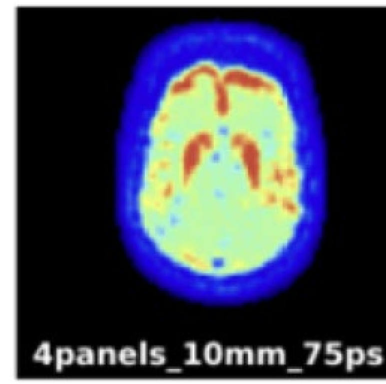
1-sided-210ps-720ps

Cherenkov-based TOF-PET with a large area MCP-PMT

Idea:
- couple short PbF_2 crystals as Cherenkov radiators to
- LAPPDs – a large area MCP-PMT, and make use of the
- flat panel concept

LAPPD with PbF_2 crystals





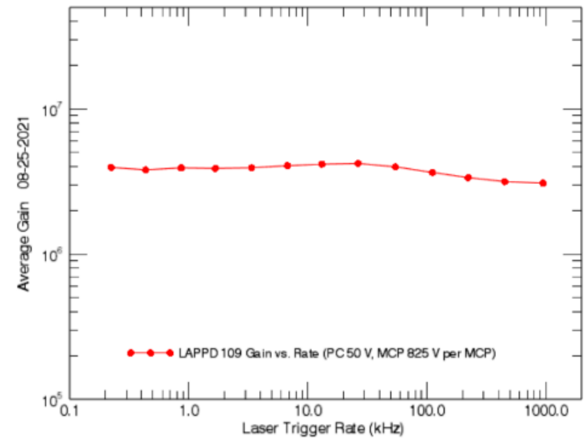
LAPPD with PbF_2 crystals attached to the entry window: an almost ideal flat panel device

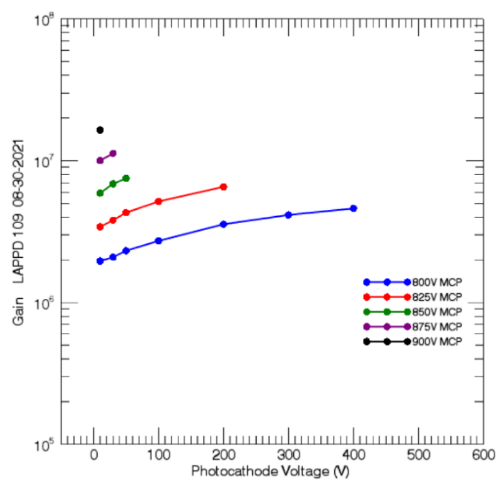
CherPET: an ERC (European Research Council) Proof-of-Principle project

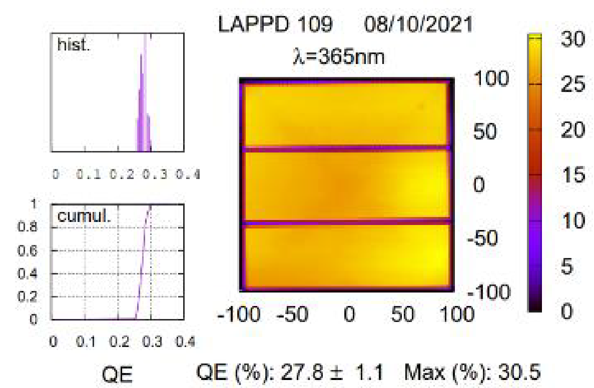
LAPPD (large area picosecond photodetector) Gen II

Characteristics (Incom):

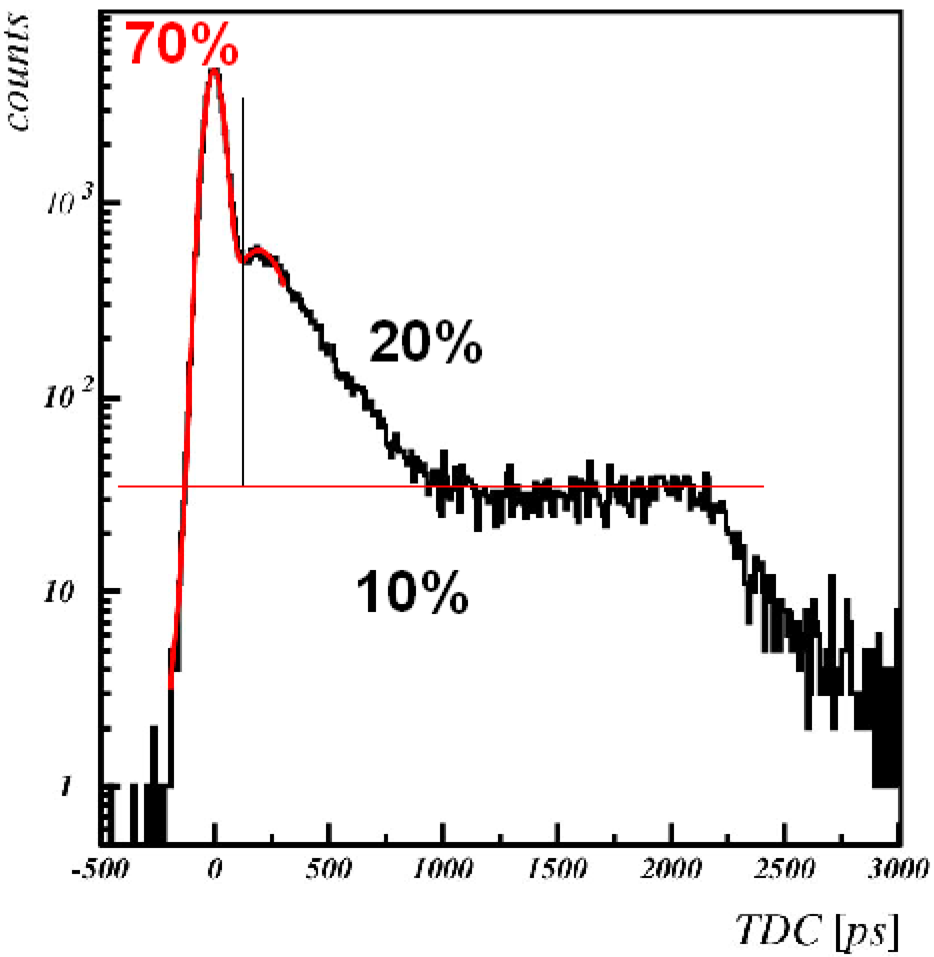
- borosilicate back plate with interior resistive ground plane anode – 5 mm thick
- capacitively coupled readout electrode
- MCPs with 20 μm pores at 20 μm pitch
- two parallel spacers (active fraction ≈ 97 %)
- gain $\approx 5 \cdot 10^6$ @ ROP (825 V/MCP, 100 V on photocathode)
- peak QE $\approx 25\%$
- size 230 mm x 220 mm x 22 mm (243 mm X 274 mm X 25.2 mm with mounting case)
- Dark Count rate @ ROP: ~ 70 kHz/cm2 with 8x10^5 gain







MCP PMT timing



Tails understood (eleastic and inelastic scattering of photoelectrons off the MCP), can be significantly reduced by:

• decreased photocathode-MCP distance and

•increased voltage difference

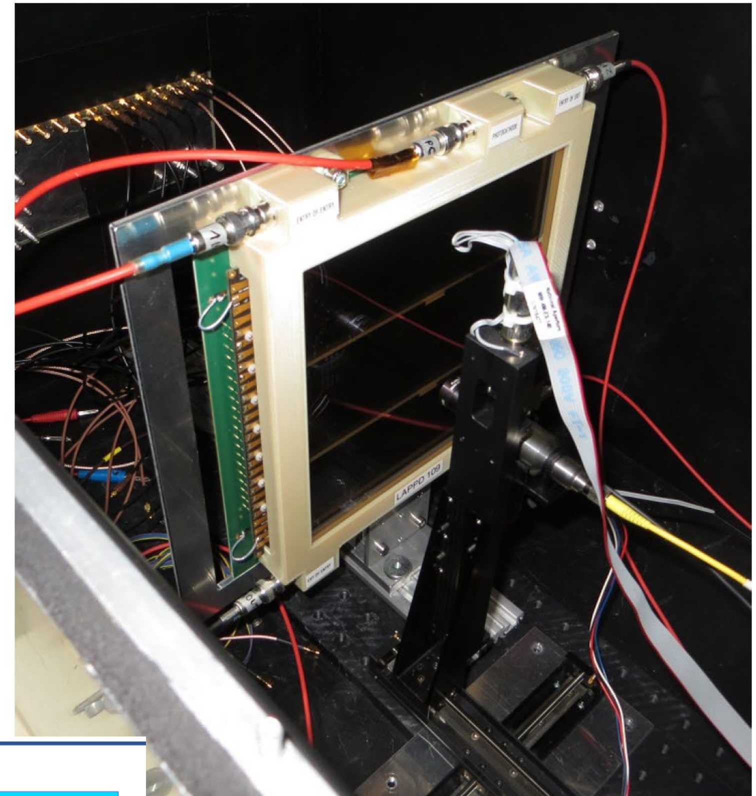
- prompt signal ~ 70%
- short delay ~ 20%
- ~ 10% uniform distribution

LPPD test set-up

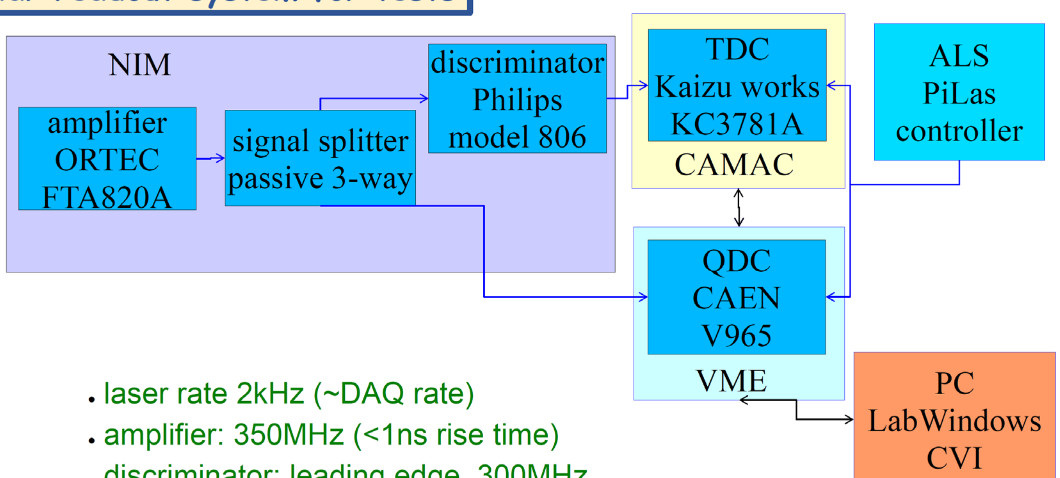
LAPPD #109:
- $\approx 200 \times 200\ mm^2$
- $20\ \mu m$ pores @ $25\ \mu m$ pitch
- resistive anode plane, capacitive coupled readout
- 5 mm thick glass backplate
- 5 HV levels: PC, MCP1in, MCP1out, MCP2in, MCP2out and resistive anode at ground potential

- Standard setup with QDC, TDC, 3D stage …
- TDC value corrected for time-walk

- ALPHALAS PICOPOWER™-LD Series of Picosecond Diode Lasers – 405 nm
- FWHM $\approx$ 20 ps
- light spot diameter < $100\ \mu m$

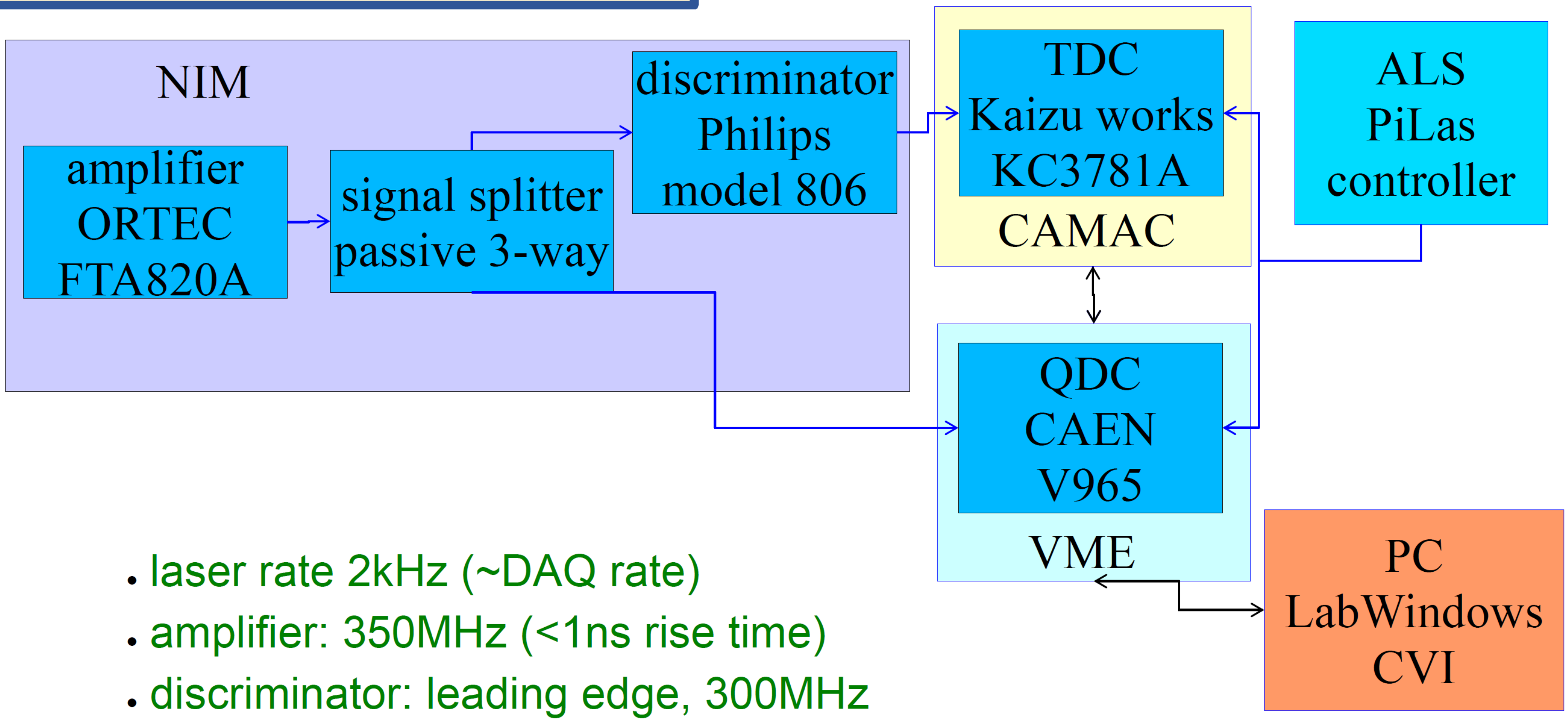


Modular readout system for tests



- laser rate 2kHz (~DAQ rate)
- amplifier: 350MHz (<1ns rise time)
- discriminator: leading edge, 300MHz
- TDC: 25ps LSB(σ~11ps)
- QDC: dual range 800pC, 200pC

Modular readout system for tests



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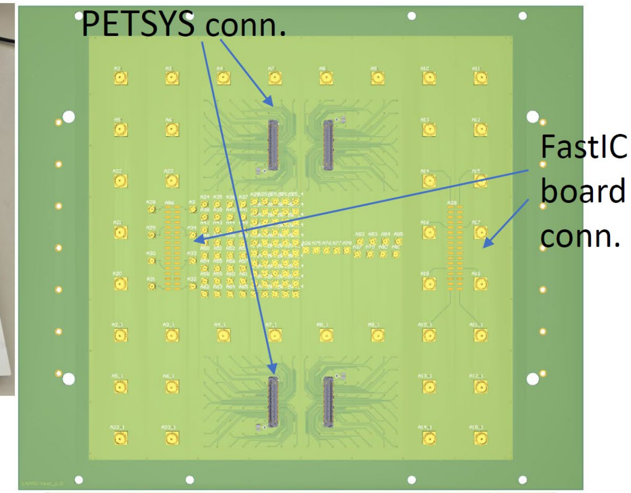
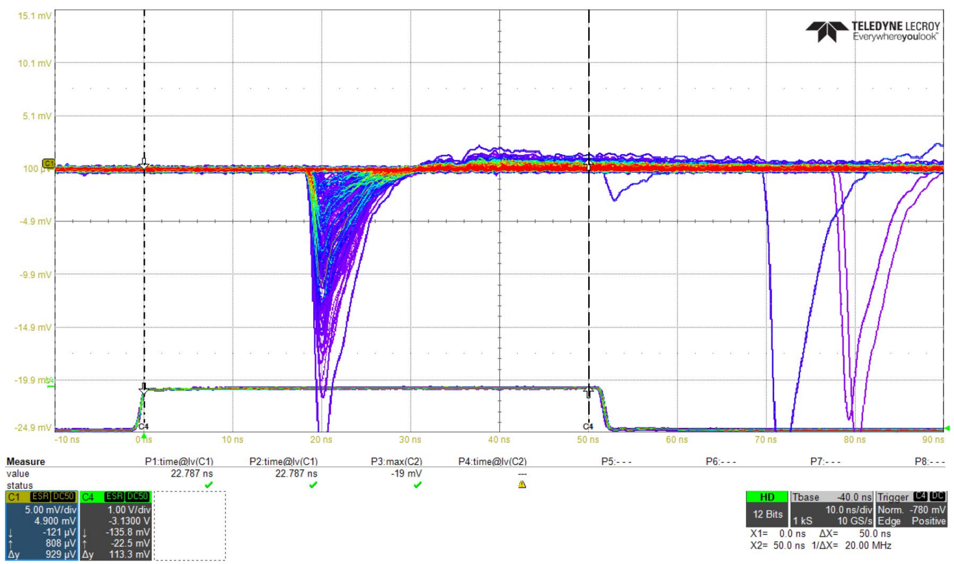
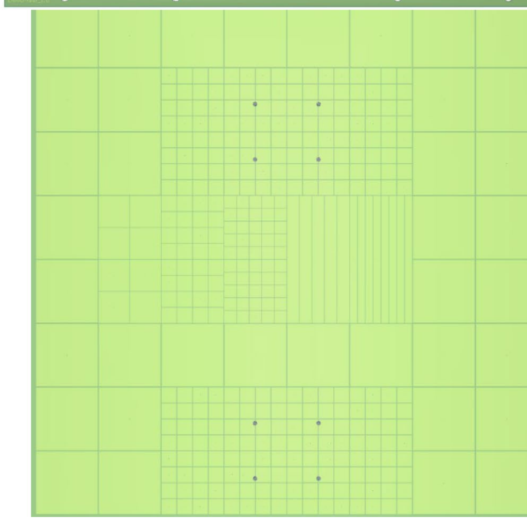
- amplifier: 350MHz (<1ns rise time)
- discriminator: leading edge, 300MHz
- TDC: 25ps LSB(σ~11ps)
- QDC: dual range 800pC, 200pC

LAPPD – IJS sensing electrodes

- capacitively coupled electrode produced at IJS with several different patterns:
  - pads: 5 mm, 6 mm, 12.5 mm, 25 mm
  - 50 mm long strips: 5 mm, 3 mm
  - PETSYS connector (256 6mm pads)
  - FastIC connector (12.5 mm and 25 mm pads)

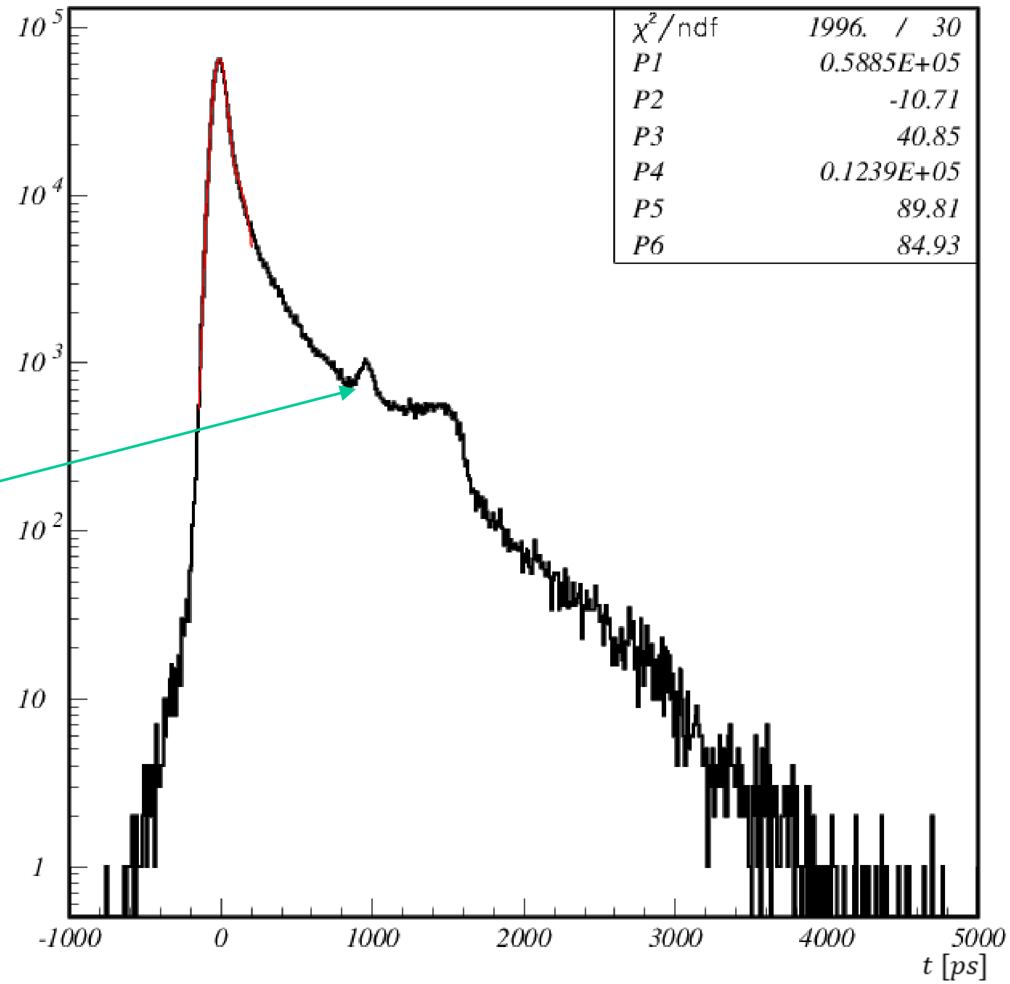






LAPPD – timing distribution

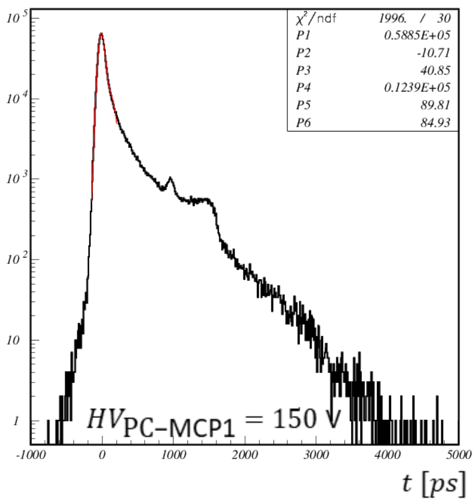
- measured timing distribution typical for MCP-PMT
- main prompt peak with some inelastic and elastic backscattering contribution
- additional small peak at about 1 ns delay probably due to some reflection (light?), delay not affected by PC-MCP1 voltage
- The plot is for the PC-MCP1 voltage of 150 V and recommended HV for others

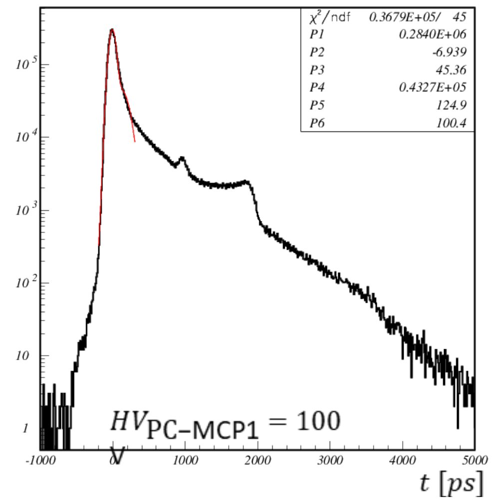


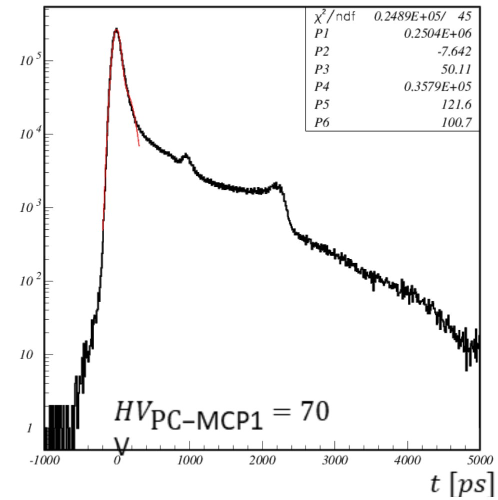
S. Korpar et al., to be submitted to NIMA

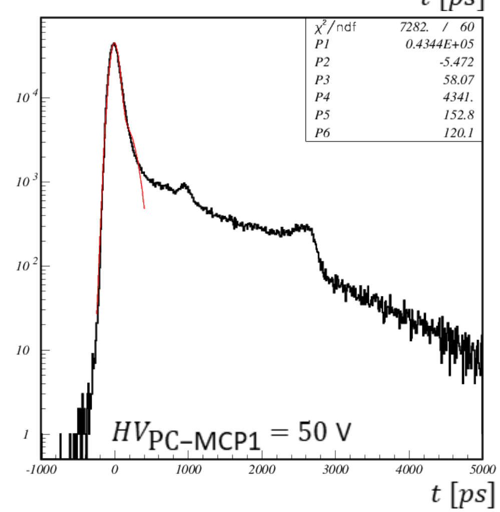
LAPPD – timing vs PC-MCP1 voltage

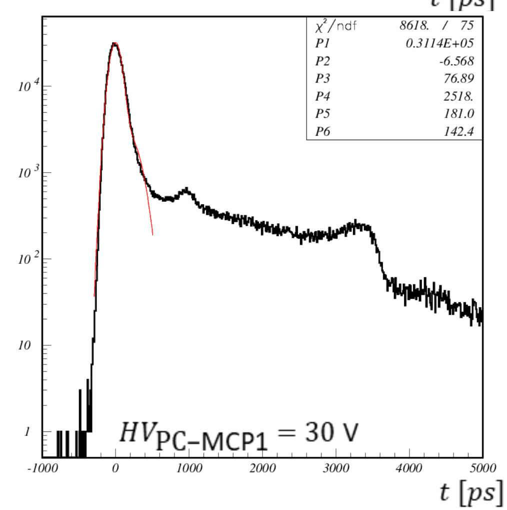
Time-walk corrected TDCs for different PC-MCP1 voltages

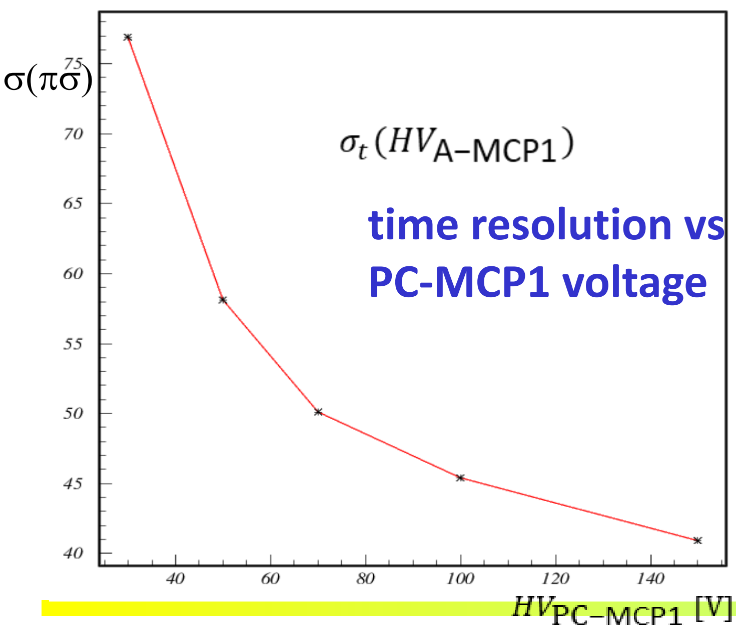






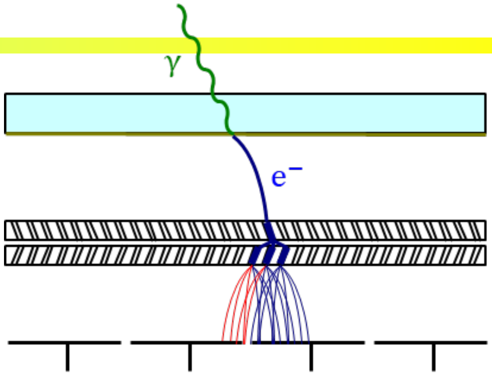




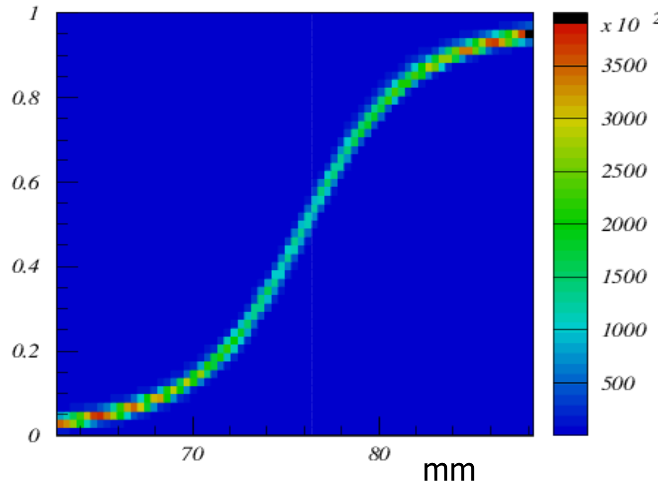


MCP PMT readout: capacitive coupling vs. internal anodes

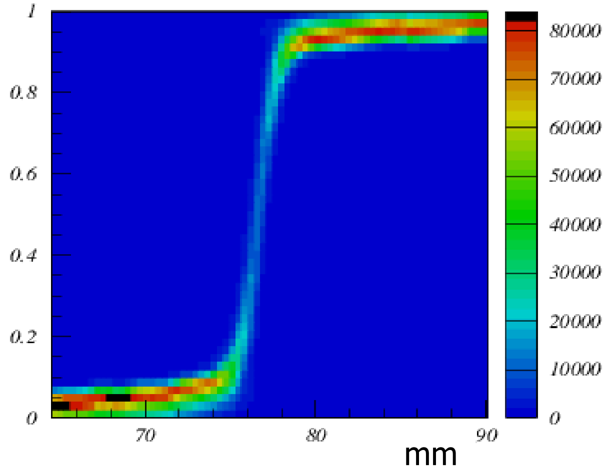
Secondary electrons spread out when traveling from the MCP-out electrode to the anode and can hit more than one anode → Charge sharing
Can be used to improve spatial resolution.



LAPPD (capacitive coupling through the backplane)



BURLE/Photonis PLANACON (internal anodes)

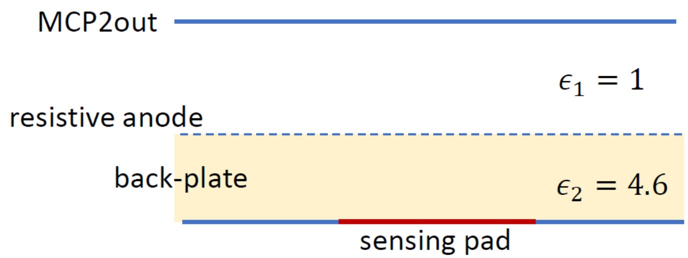


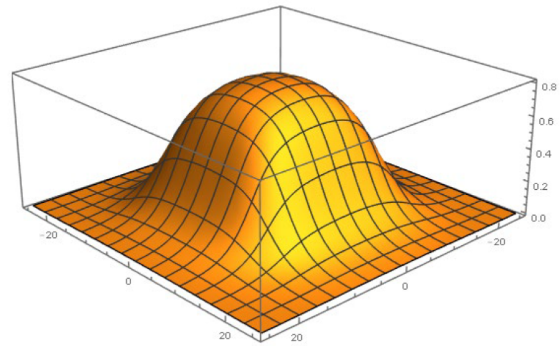
Fraction of the charge detected by the right pad as a function of red laser spot position

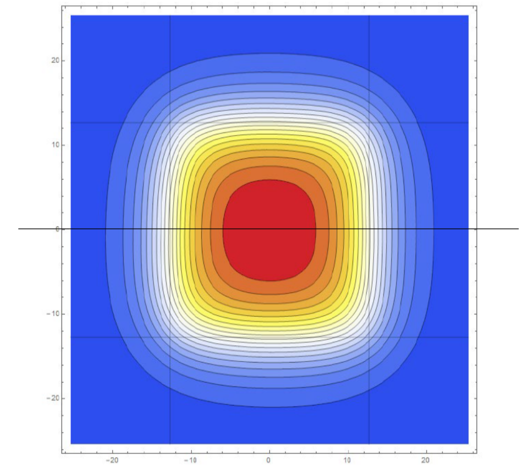
Capacitive coupling vs. internal anodes: signal spread comparison for two MCP PMTs with the same pad size, same range: charge sharing is more effective for capacitive coupling (spreads over larger area) - advantage or not: depends on the usage

LAPPD charge sharing

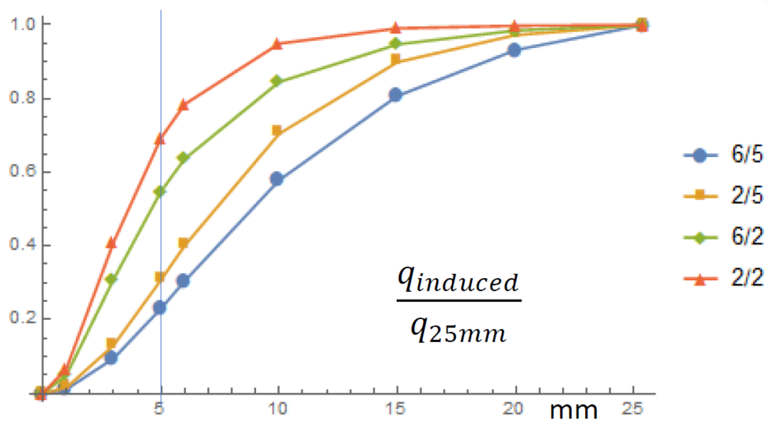
- calculation of charge sharing for different MCP2out-resistive anode/resistive anode-sensing electrode distances (6/5-measured, 2/5, 6/2, 2/2)

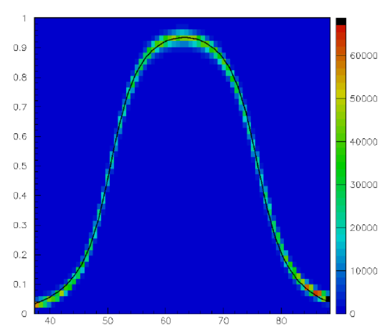


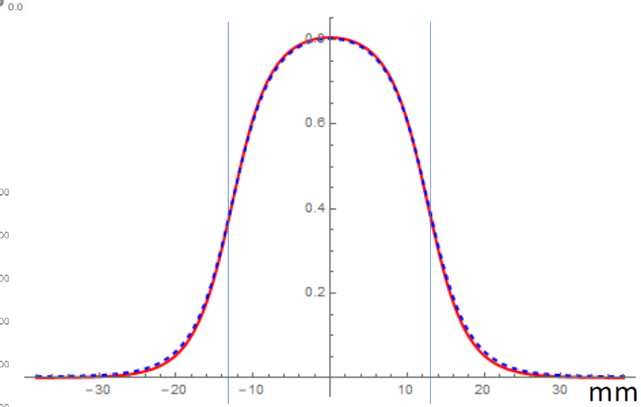




- fraction of the charge induced vs. square pad size when signal is produced in the centre of the pad







- Nice agreement between modeling and measurements.
- For a better spatial resolution, a thinner back-plate would be needed.

Summary

Interplay of detector R&D for particle physics and medical imaging has a long history, and this will remain one of the sources of innovation in medical imaging

Limited angle devices with very fast gamma detection look very promissing – lower cost, flexibility in use, affordable total-body scanner

Cherenkov radiation based annihilation gamma detectors offer a promising method for very fast detection and potentially cheaper devices