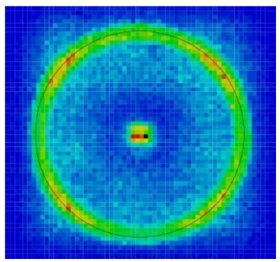
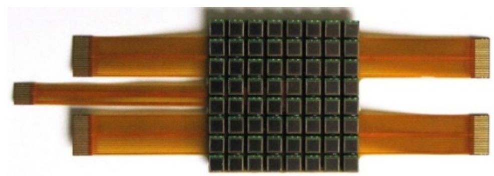
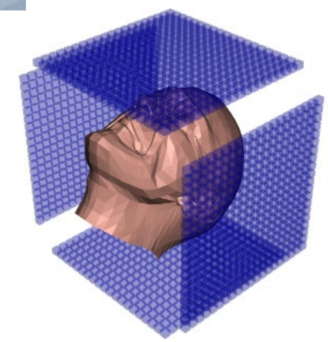
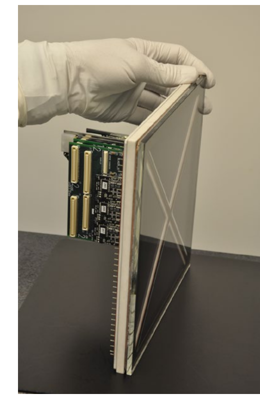








Instrumentation for advances in PET medical imaging

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, can often be applied in medical imaging

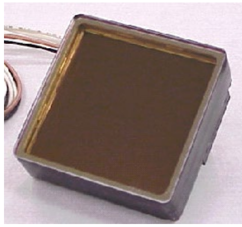
… and sometimes also vice versa…

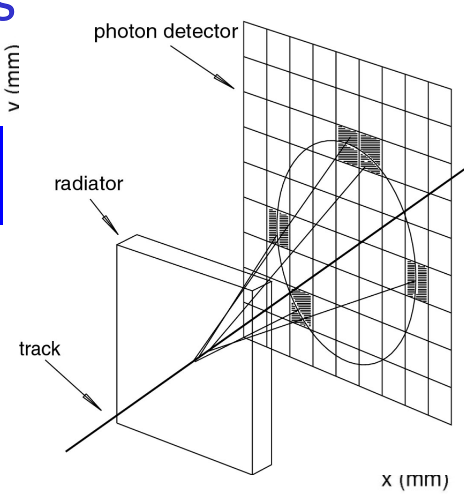
One of the recent examples: SiPMs as scintillation light sensors for

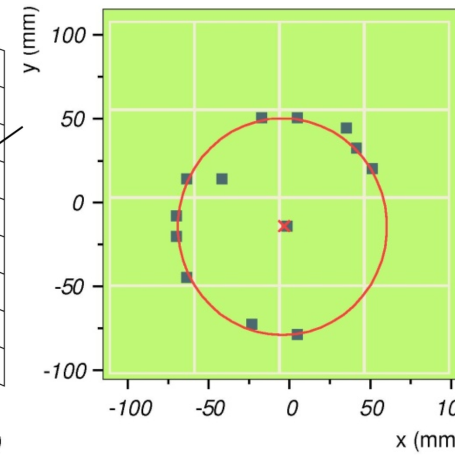
- Electromagnetic calorimeters, RICH counters, scintillating-fiber trackers
- Positron Emission Tomography (PET) scanners

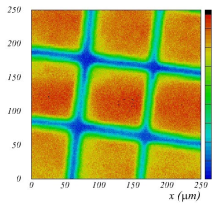
Also: in HEP we are often small users, it helps us if a device becomes interesting for big users.

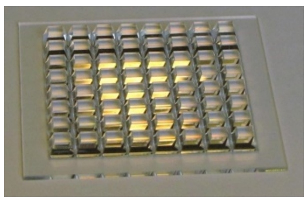
Our original expertise in instrumentation: Cherenkov detectors, single-photon sensors and associated electronics

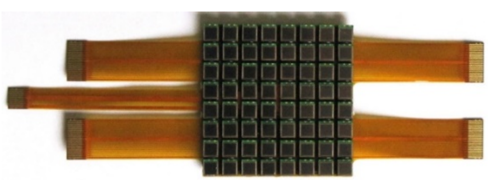


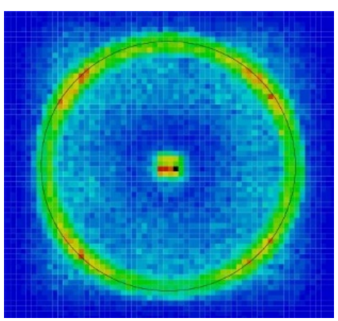


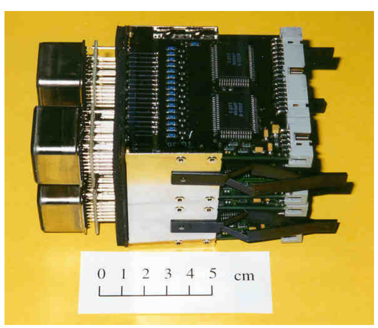


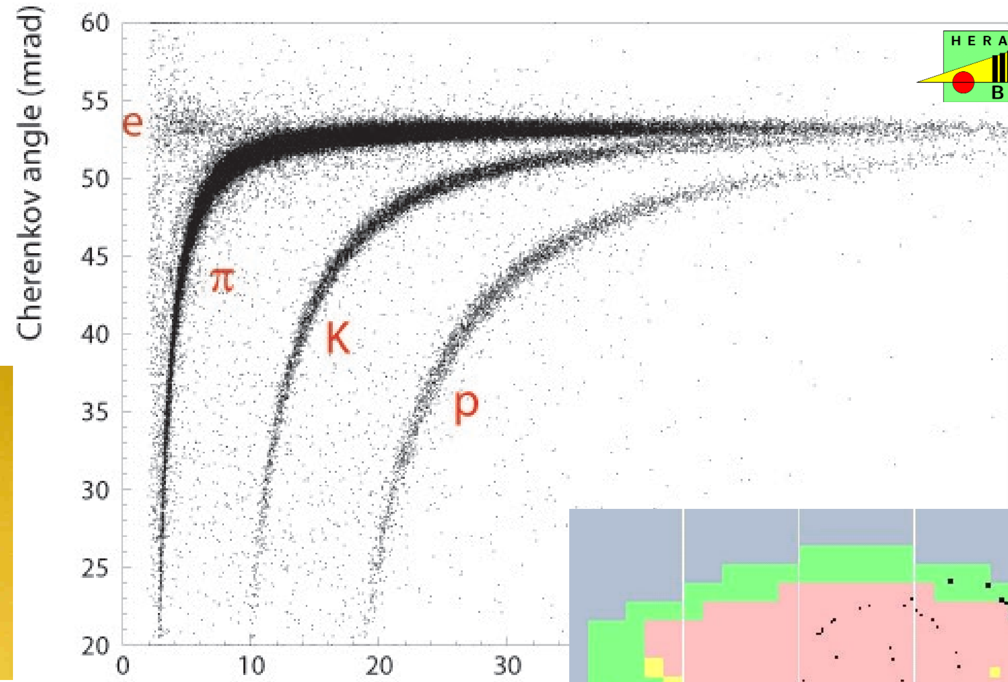


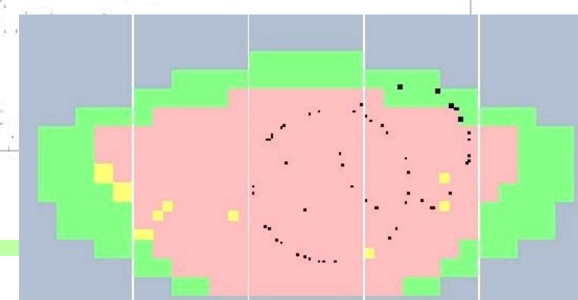




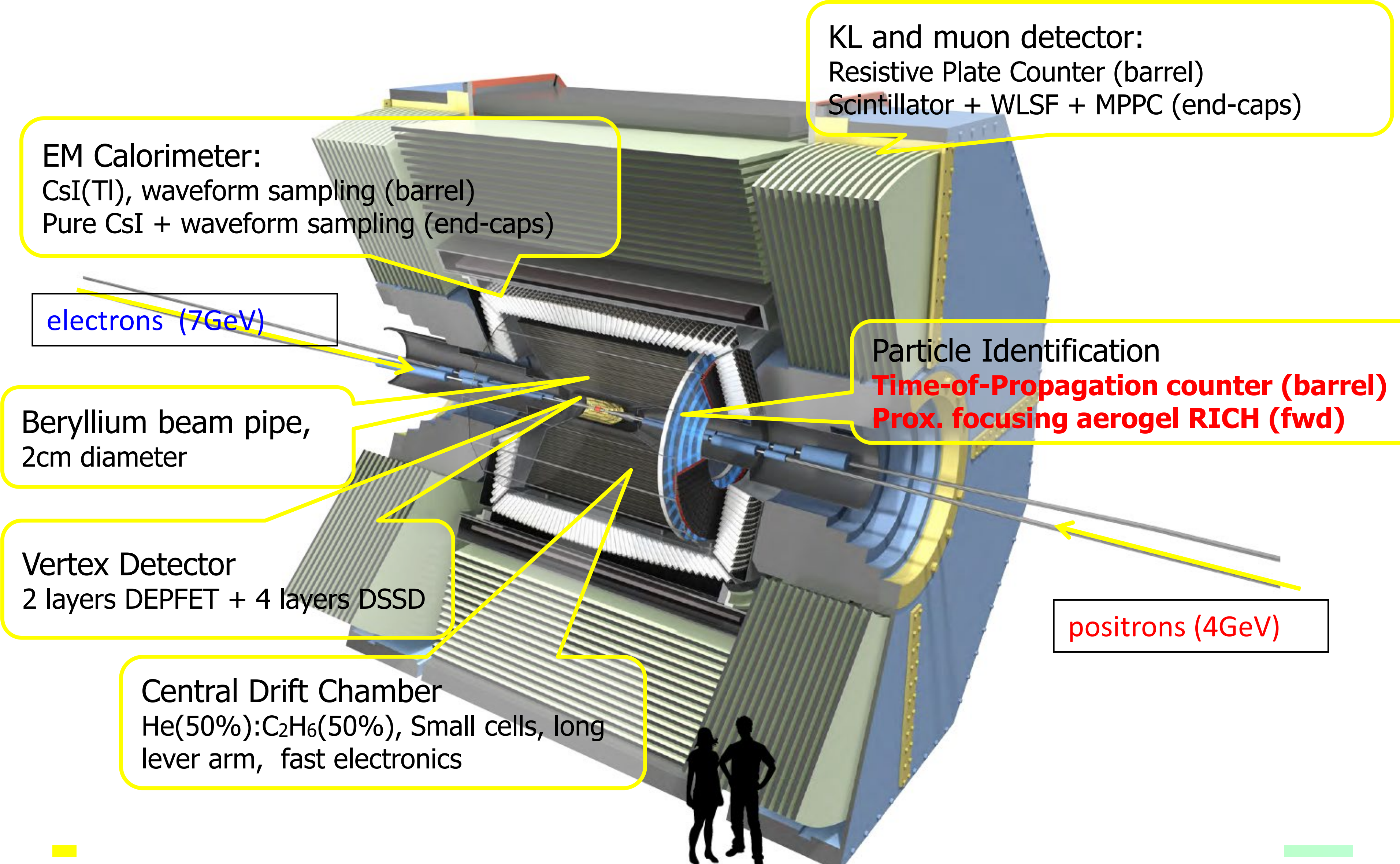








Belle II Detector

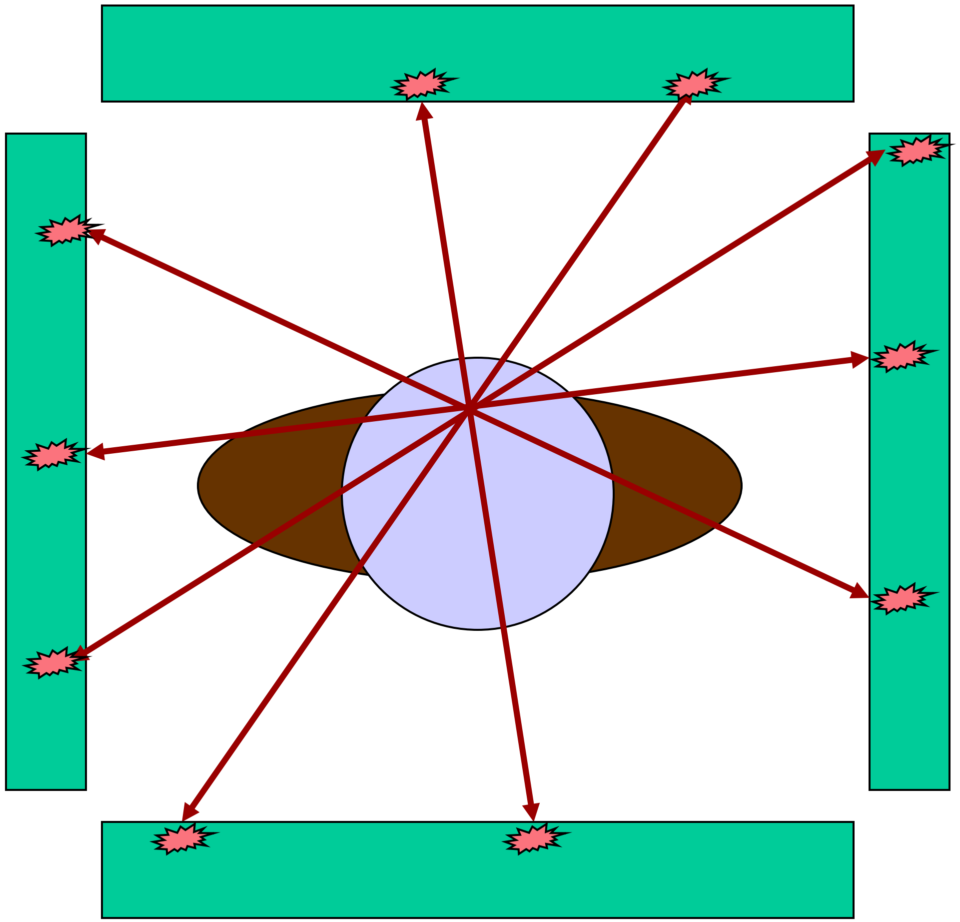


Contents

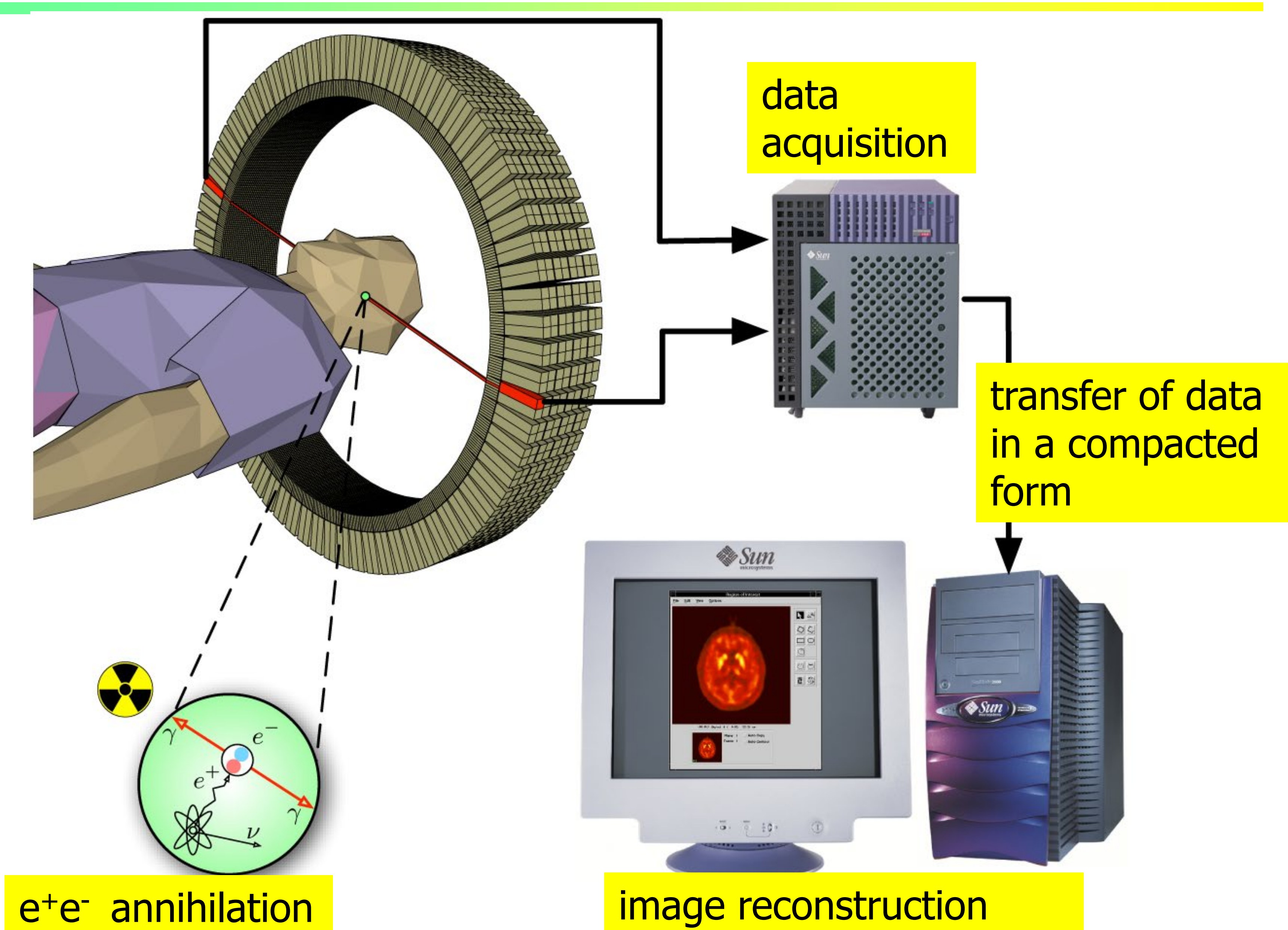
PET: positron emission tomography

In the blood of the patient, a substance is administered that contains a radioactive isotope – a beta+ emitter (e.g., fluorodeoxyglucose, FDG, with ^{18}F).

Positrons from the ^{18}F decay annihilate with electrons in the tissue, emitting a pair of collinear gammas.



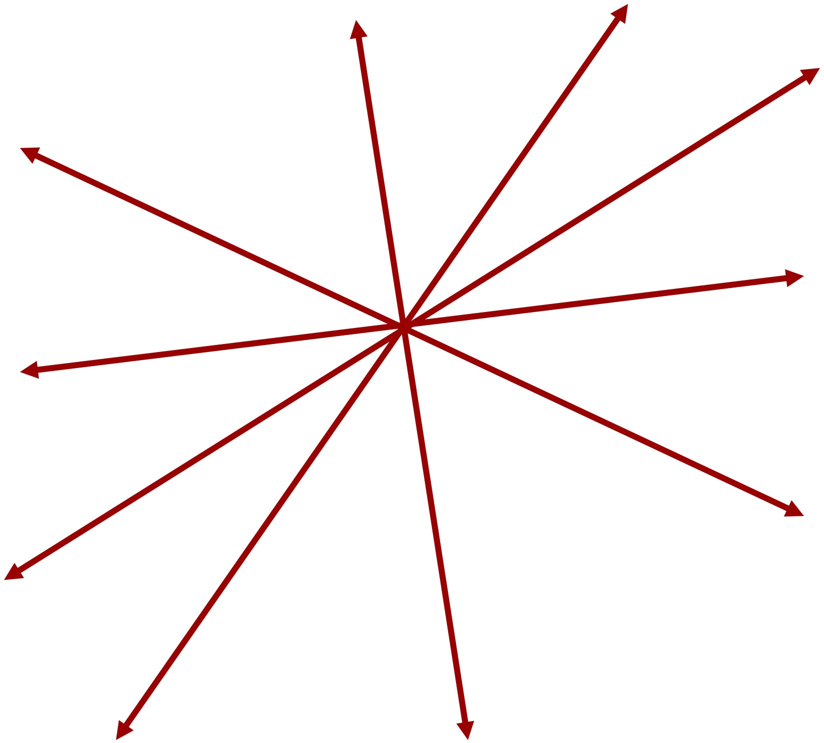
PET: collection and handling of data

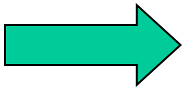


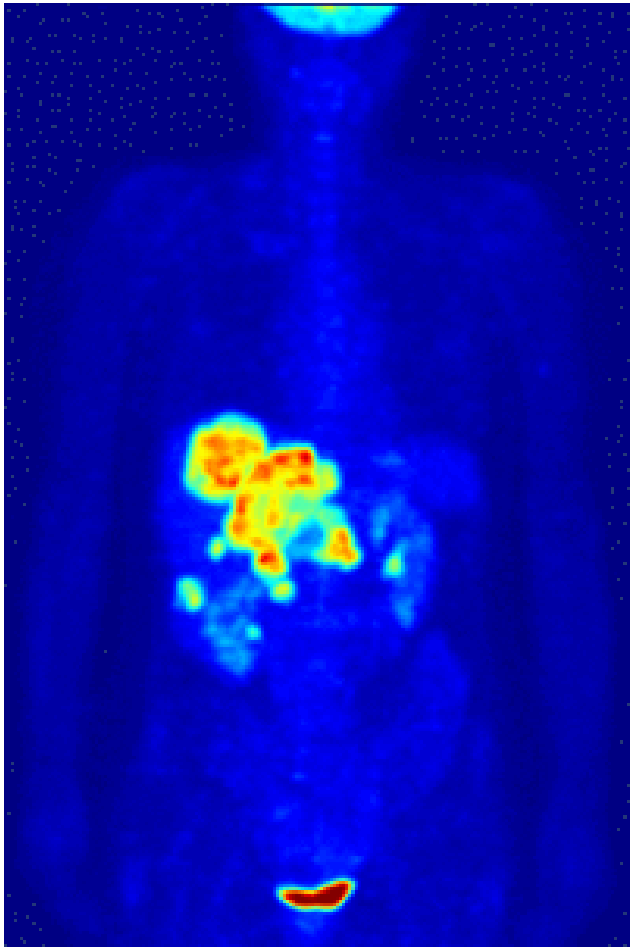
PET: image reconstruction

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

The places in the body with a higher substance concentration will show a higher activity.







PET with a time-of-flight information

The emission point of the γ pair can be anywhere on the line between the two detector elements that have been hit.

Detectors can in principle also measure the time of arrival of each of the γ rays

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

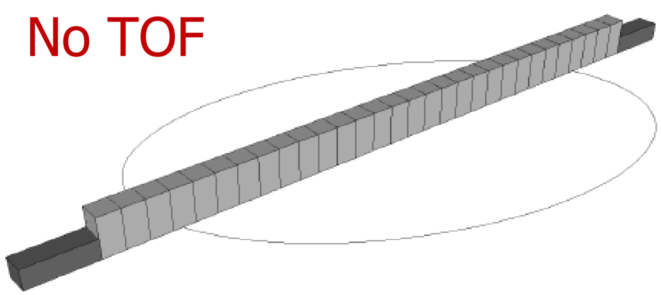
→ time-of-flight (TOF) PET

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

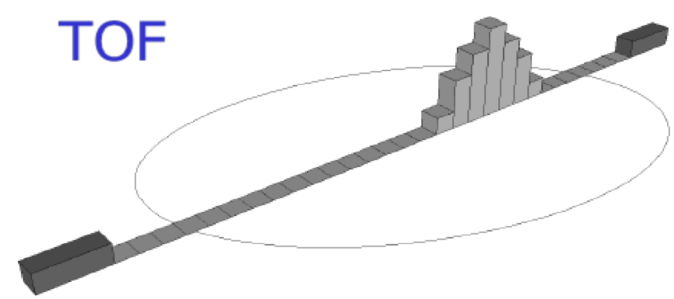
In the reconstruction step, each line contributes to fewer pixels

→ less noise in the reconstructed image

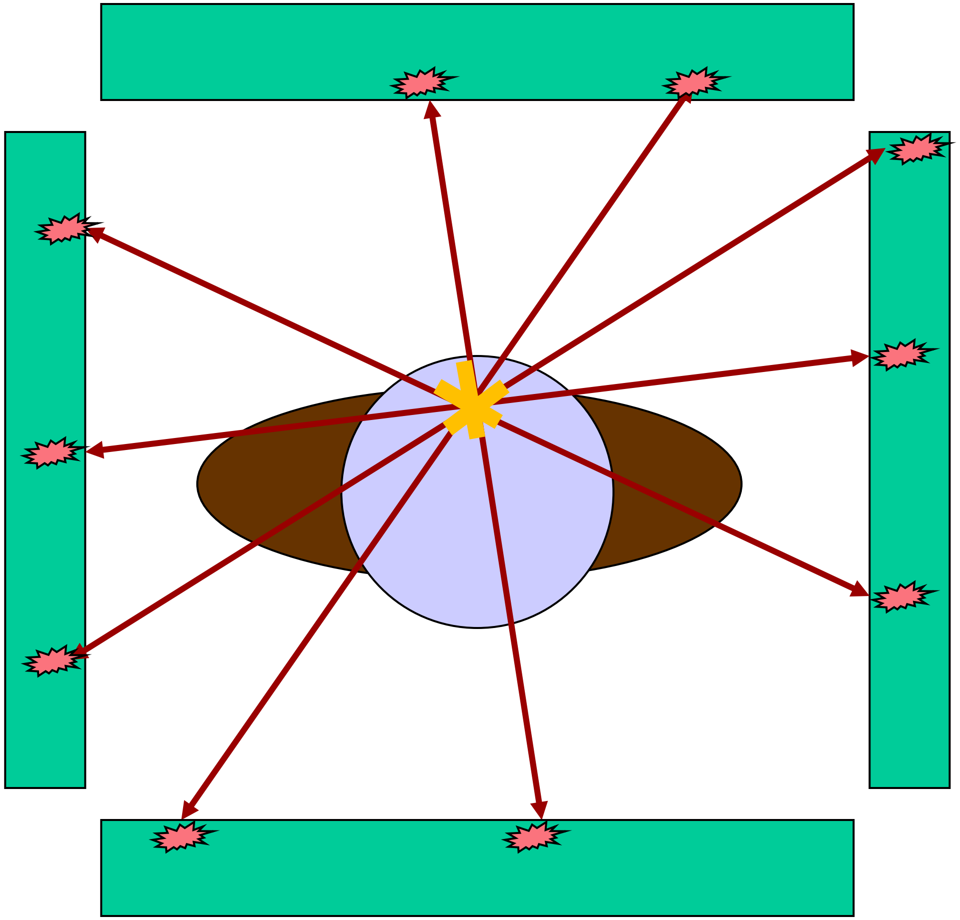
No TOF



TOF



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



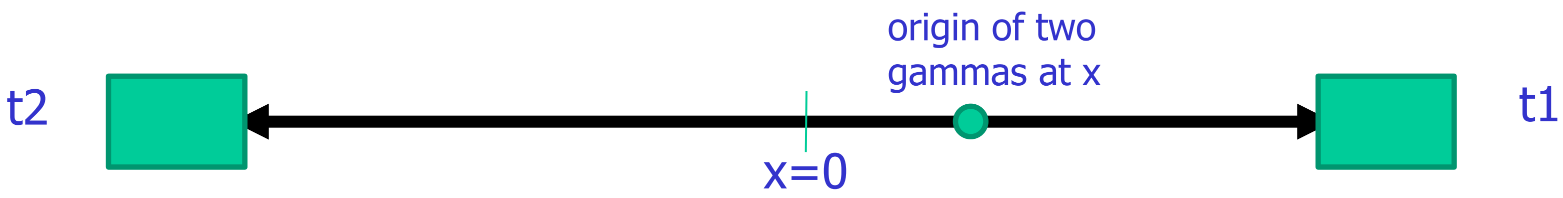
Good resolution in time-of-flight → limits the number of hit pixels along the line connecting the two detector hits

In the reconstruction step, each line contributes to fewer pixels

→ less noise in the reconstructed image

TOF-PET: time resolution

What kind of time resolution is needed?



t1 = (L/2 – x)/c   source at x, distance between detectors = L

t2 = (L/2 + x)/c

t1 – t2 = 2x/c → 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) – CTR, coincidence timing resolution (FWHM)

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

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

Optimize detector CTR (Δt) to maximize the sensitivity

Motivation for Fast TOF PET

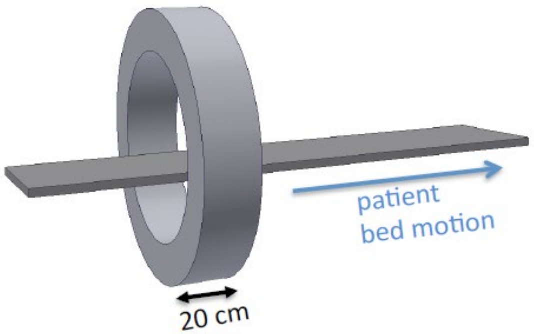
- Paradigm shift in medicine:
  - From the treatment of an obvious disease
  - To early diagnosis / prevention
- This leads to more stringent requirements on PET diagnostics
  - Sensitivity (=positive→positive)
  - Specificity (=negative→negative)
- Targeted Radionuclide Therapy (TRT) & Theranostics*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

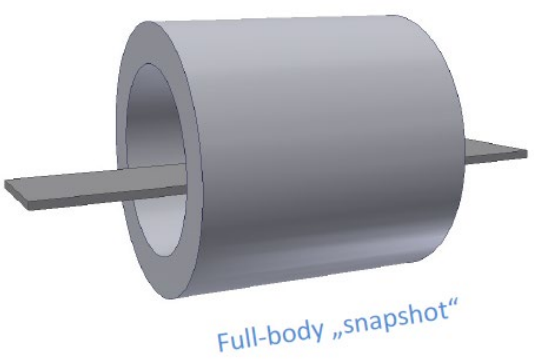
Number of PET scanners per million people

*Theranostics is a two-pronged approach to diagnosing and treating cancers through the use of radiotracers. Radiotracers are compounds made of chemicals that selectively bind to a specific target in the body, and of a radiative component. In the diagnostic phase, the radioactive part is a beta emitter, while in the treatment phase it is a strong radiation source to damage the cancer cells.

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 resolution

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

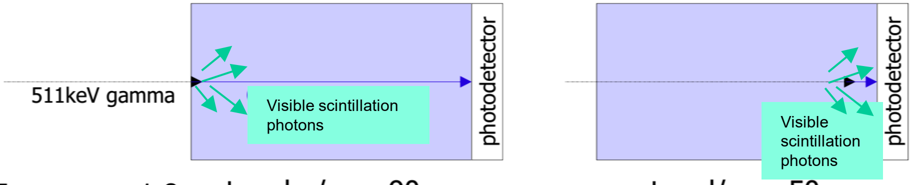
Photodetector
- converts optical photons into electrical pulses

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**

Limitations on timing due to optical travel time

Inherent limitation for any crystal-based annihilation gamma detector:

- 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 → $\Delta t = 40$ ps

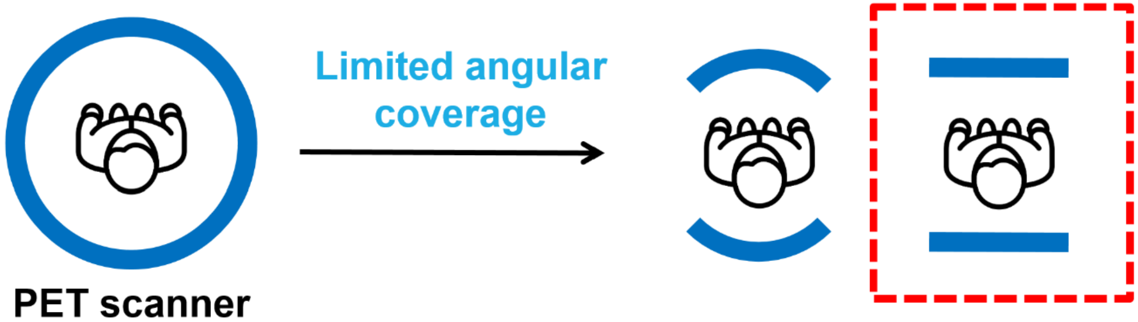
- Can in principle be corrected for by:
  - measuring the depth of interaction (DOI)
  - building the detector with shorter crystals → multi-layer configuration

Can we simplify the TOF PET scanner

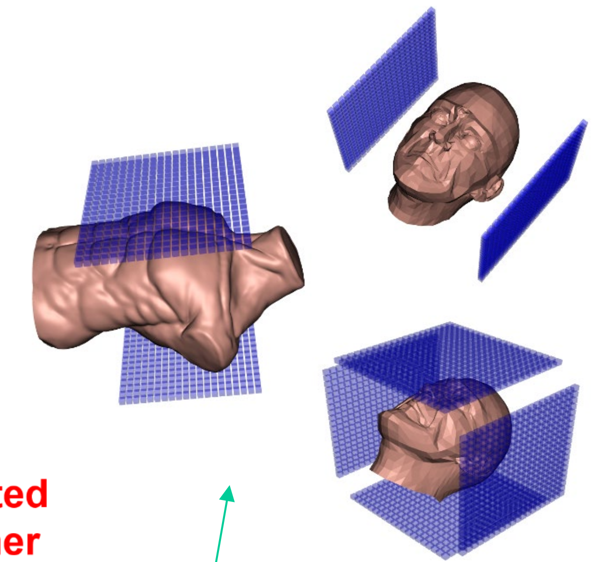
– and make it cheaper and flexible?

Next generation scalable time-of-flight PET

Superb time resolution enables simplifications in the scanner design



Panel-based limited angle PET scanner



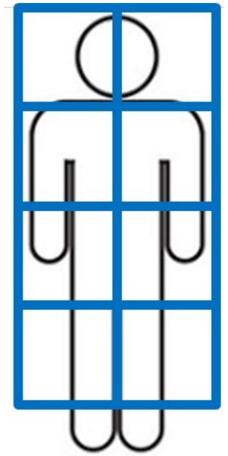
Limited angle PET scanners will generally produce distorted images with artefacts - unless they have good **time-of-flight** information

S. Surti, J. S. Karp,
Phys. Med. Biol. 53 (2008) 2911

G. Razdevšek *et al.*,
IEEE TRPMS 6 (2022) 721

Potential benefits of a panel-based PET system

- 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



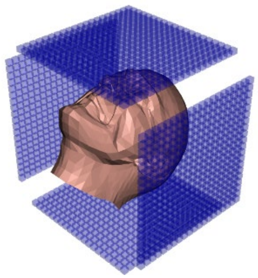
Simulation of a limited angle system

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

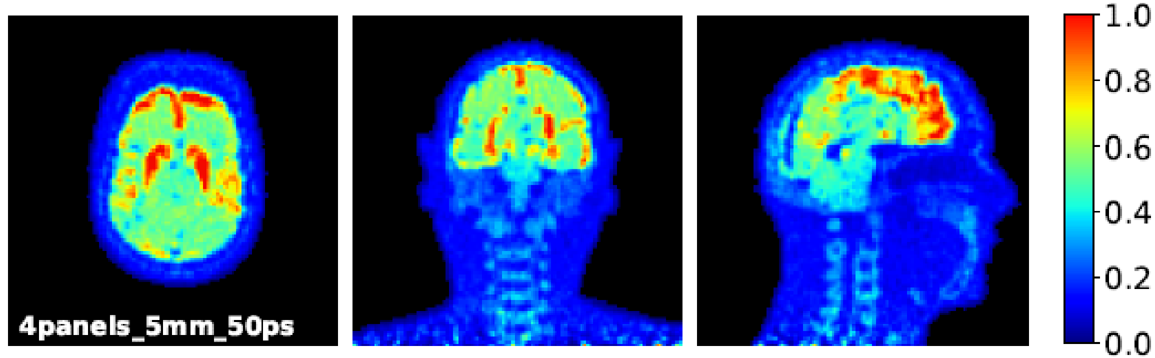
CASToR → image reconstruction with Maximum Likelihood Expectation Maximization (**MLEM**) algorithm

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

Enabling Open Geometry systems

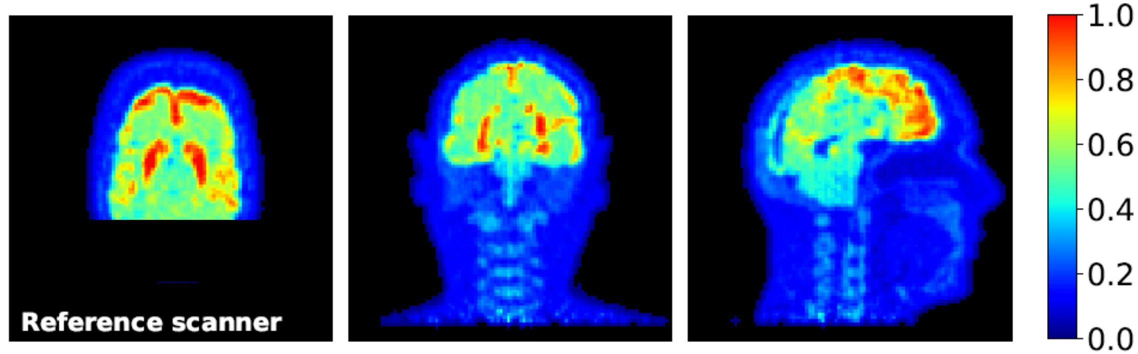


Two pairs of 30x30 cm^2
LYSO panels with 50ps CRT



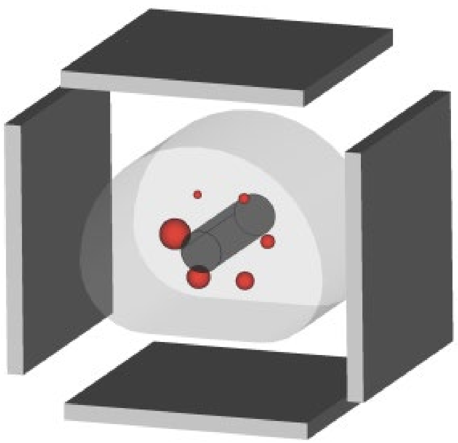
Similar performance as

Siemens Biograph Vision

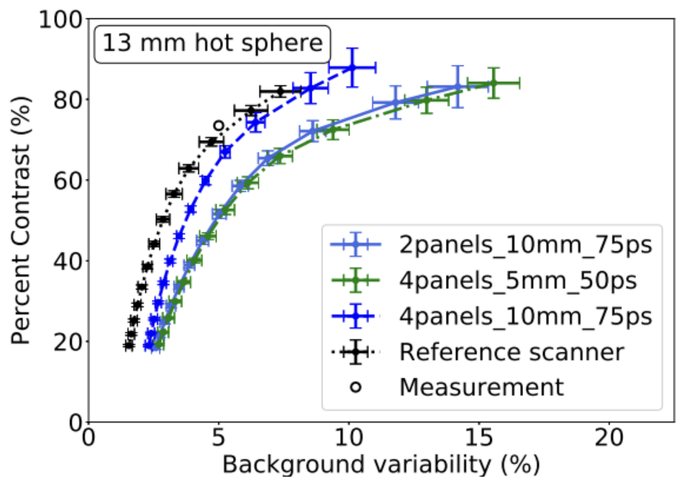


G. Razdevšek *et al.*, "Multi-panel limited angle PET system with 50 ps FWHM coincidence time resolution: a simulation study," IEEE TRPMS 6 (2022) 721, doi: 10.1109/TRPMS.2021.3115704.

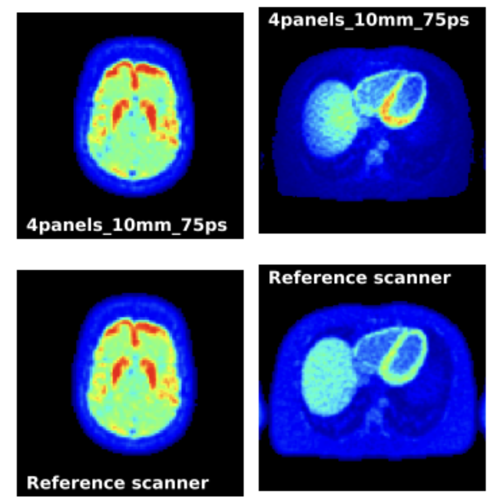
Simulation study of planar configurations



Simulated arrangement of 30x30 cm^2 flat panel detectors



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



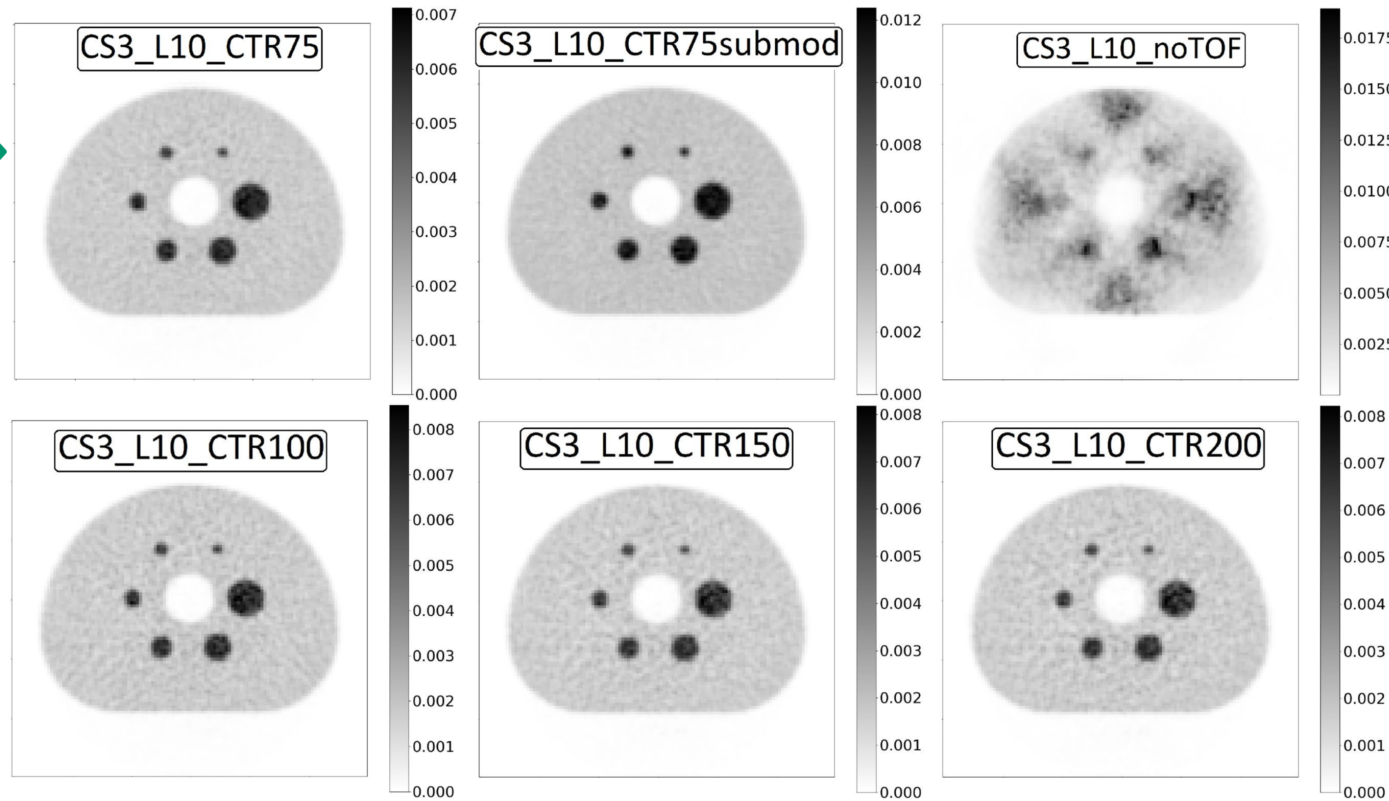
Reconstructed images of a torso and head for the flat panel detectors and the reference scanner Siemens BV

G. Razdevšek *et al.*, IEEE TRPMS 6 (2022) 721

Design optimisation of a flat-panel, limited-angle TOF-PET scanner: simulation

Reconstructed NEMA phantom for various detector parameters.

Scanner: 3mm x 3mm x 10mm,
CTR=75ps



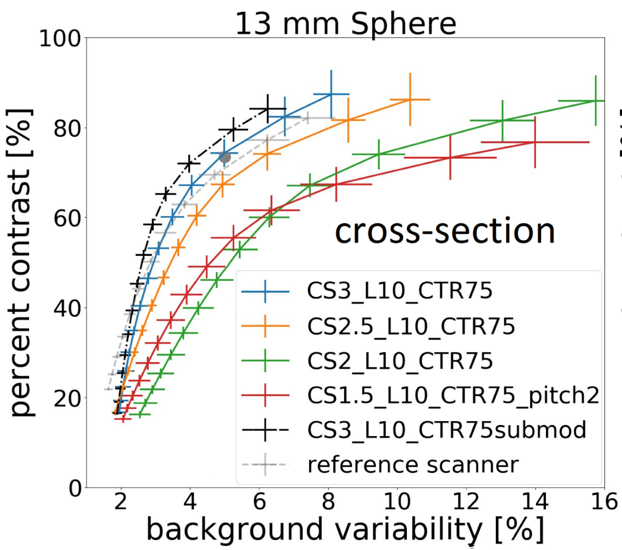
A 5 mm filter is applied to all of the images.

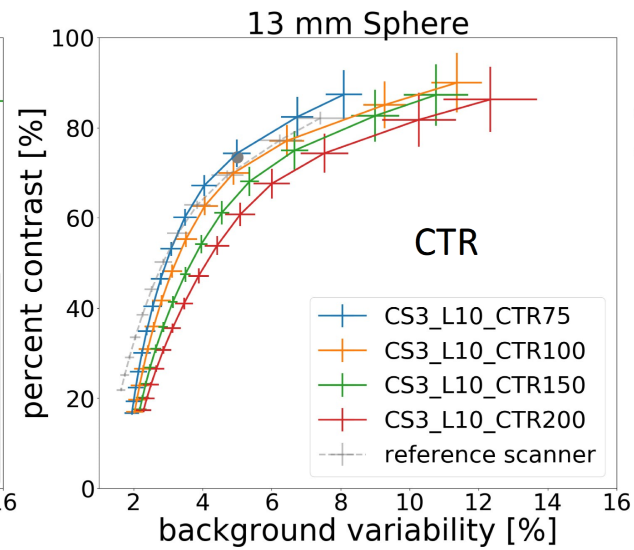
M. Orehar et al., Design Optimisation of a Flat-Panel, Limited-Angle TOF-PET Scanner: A Simulation Study. Diagnostics 14 (2024) 1976, doi: 10.3390/diagnostics14171976.

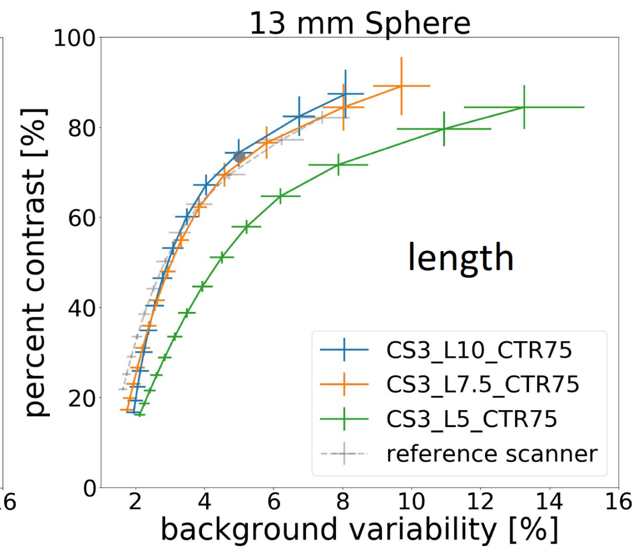
Design optimisation of a flat-panel, limited-angle TOF-PET scanner: simulation

Image quality plots for the 13 mm sphere for different scanner parameters.

Percent contrast vs background variability (~noise)







different crystal cross-sections and pitches

Impact of CTR

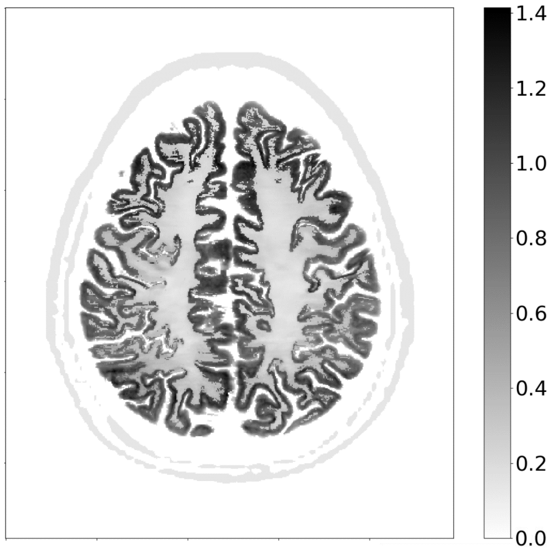
Impact of crystal length.

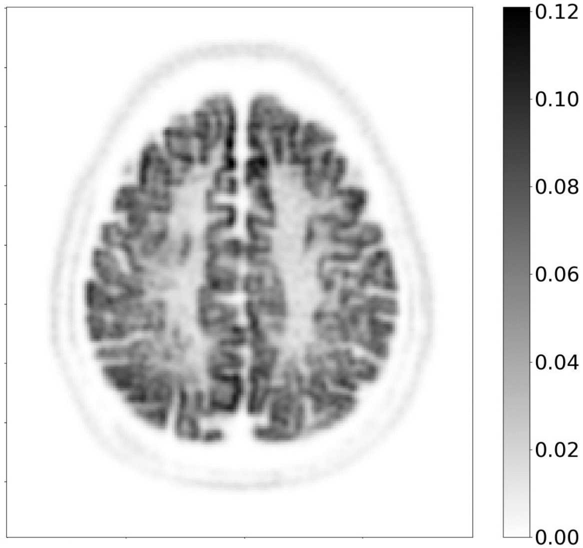
Grey circle: experimental measurement value for the reference scanner.

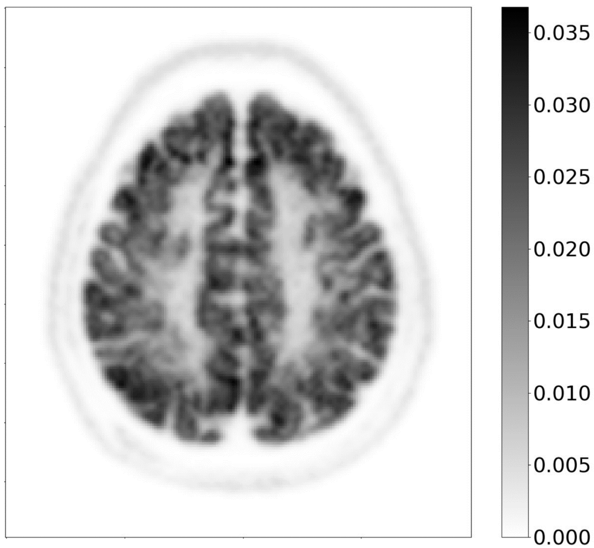
M. Orehar et al., Diagnostics 14 (2024) 1976

Design optimisation of a flat-panel, limited-angle TOF-PET scanner: simulation

Images of the brain phantom in the transverse plane







activity phantom

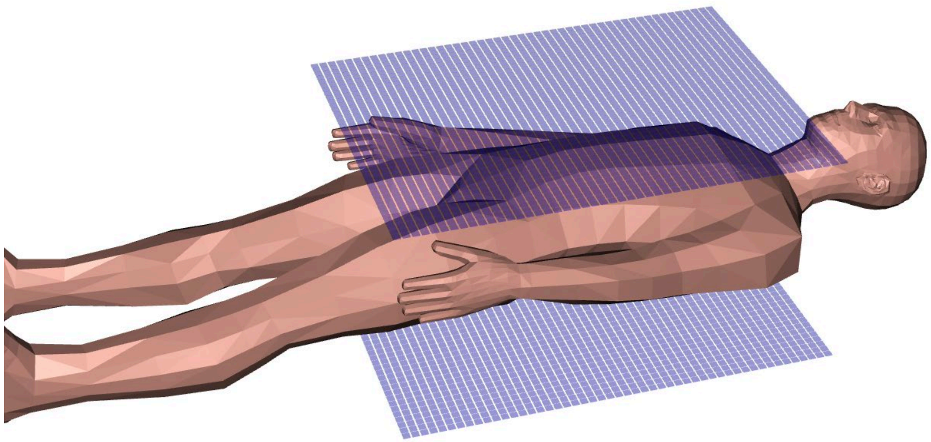
image reconstructed with the 3mm x 3mm x 10mm, CTR 75ps scanner

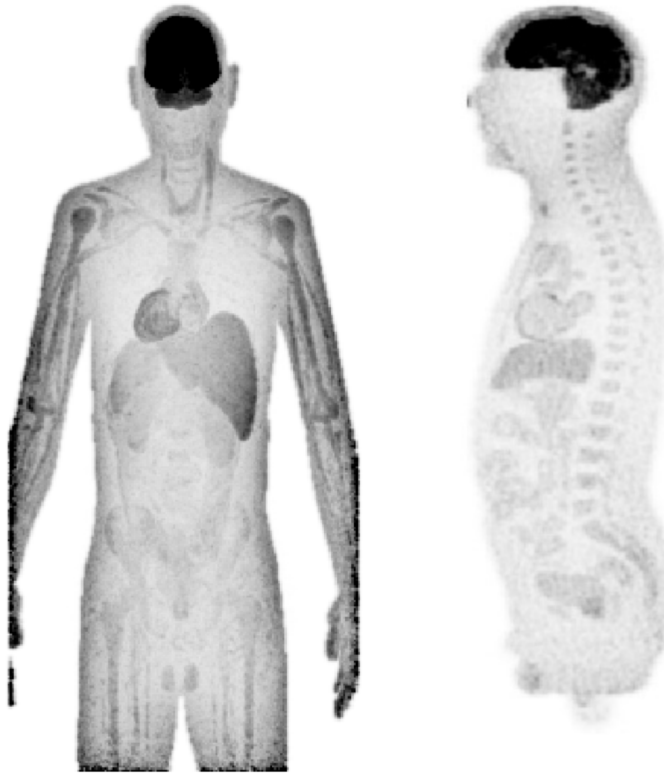
image reconstructed with the reference scanner.

M. Orehar et al., Diagnostics 14 (2024) 1976

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 a 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.

Next-generation scalable time-of-flight PET

Address PET system challenges of a limited angular 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

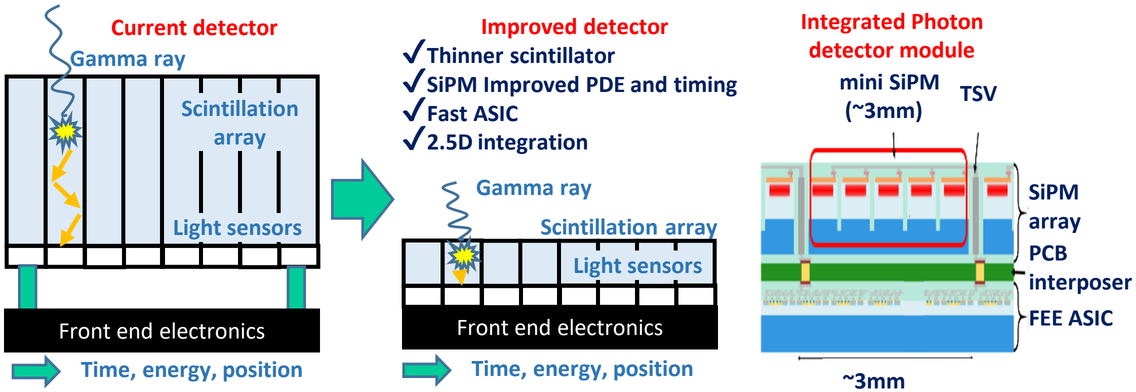
Managed to get a 3 MEUR EIC EU grant for 5 years to further develop the method and construct a prototype.
PI: Rok Pestotnik (JSI) https://petvision.org

PET VISION

Supplemented by a recently awarded NIH grant to Yale.

Fast CTR PET module

How to achieve such a good CTR?



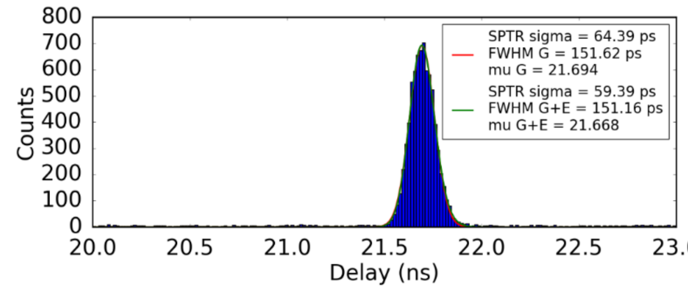
- Sensor: SiPM development at FBK
- FEE ASIC: FastIC, 8-channel ASIC for fast single photon sensors, a collaboration of ICCUB (Univ. Barcelona) and CERN

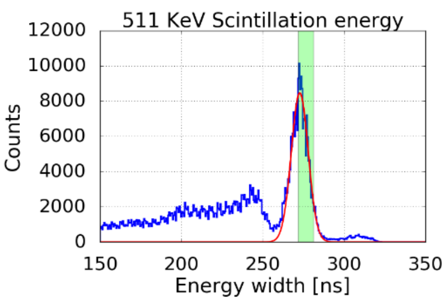
First results with FastIC

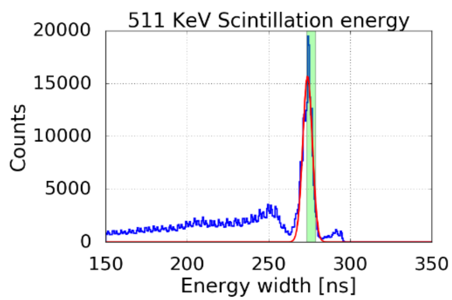
- **Sensor:** FBK-NUVHDLFv2b 3x3 mm², 40 pixel pitch.
- **Crystal:** LSO:Ce Ca 0.2% of 2x2x3 mm³.

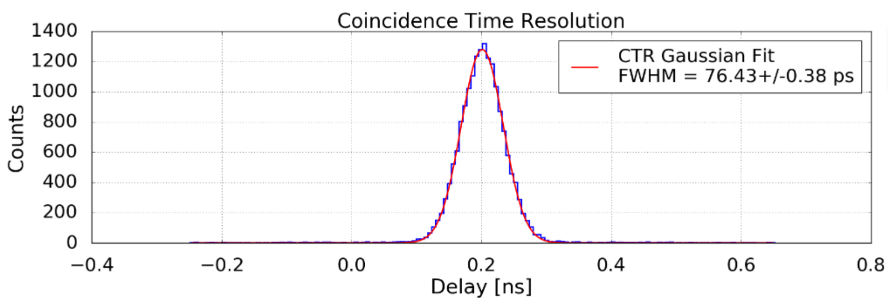
Single photons

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









FWHM = 76.43 ps

Pairs of annihilation gammas

D. Gascon, talk at Instrumentation for the future of particle, nuclear and astroparticle physics and medical applications in Spain, March 2023

- ICCUB and CERN are working on FastIC+: integration of 25 ps bin TDC on FastIC
- On the longer term plan for a 32 ch. ASIC (FastIC32)

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

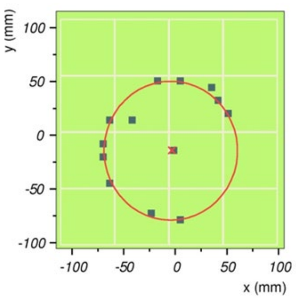
R. Dolenec[a,b], D. Consuegra Rodríguez[a], P. Križan[a,b], M. Orehar[b], R. Pestotnik[a], G. Razdevšek[b], A. Seljak[a] and S. Korpar[a,c]

[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/

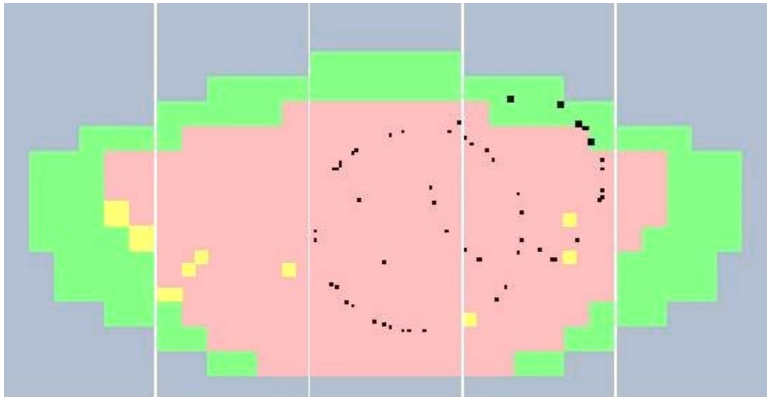
Imaging Cherenkov detectors

Measure the Cherenkov angle (RICH counter)

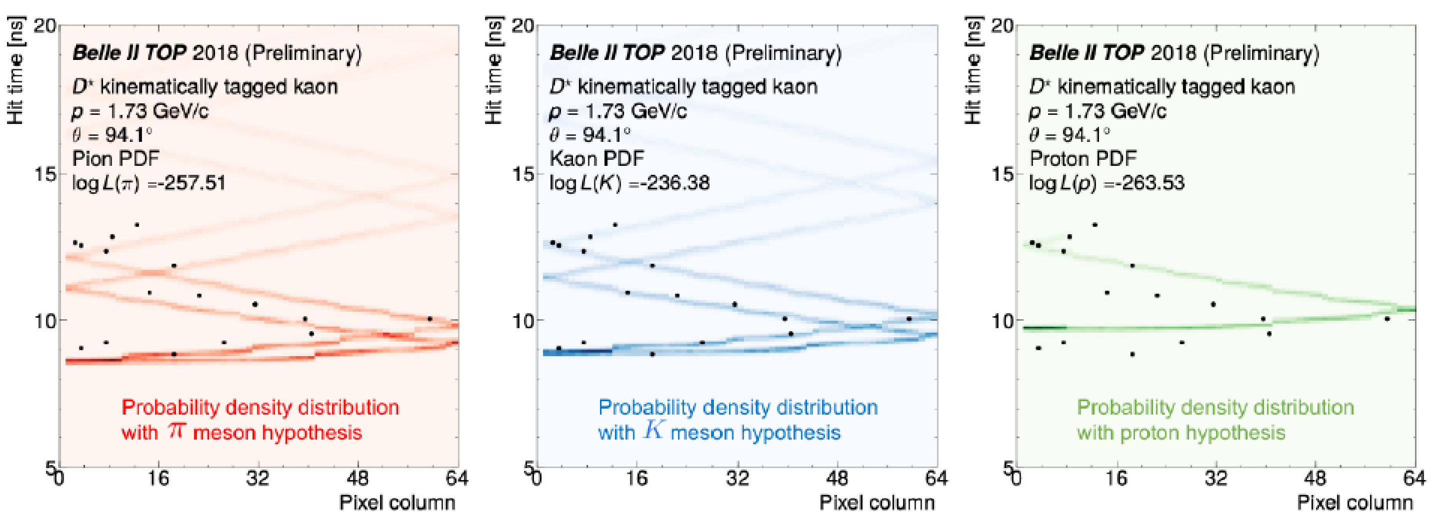


ARICH@Belle II

RICH@HERA-B

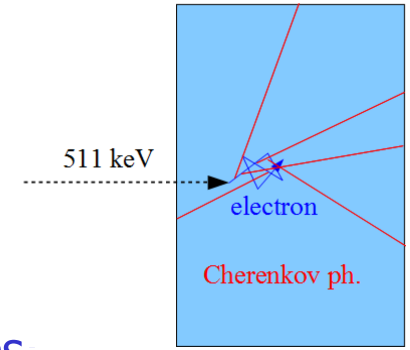


… or a pattern in the coordinate-vs-time space (TOP@Belle II)



Use of Cherenkov Light in TOF-PET

γ detectors in traditional PET: scintillator crystal + photodetector



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

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

	BGO	LSO	PbF_2
Density (g/cm³)	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)

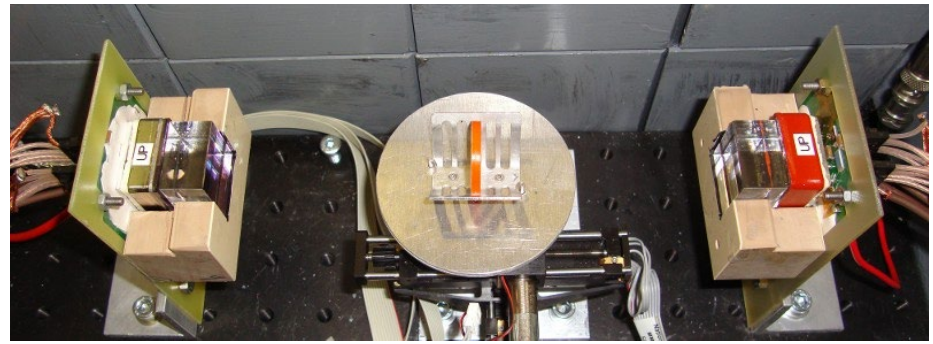
- low price (**1/3 BGO, 1/9 LSO**)

Mao, IEEE TNS 57:6 (2010) 3841

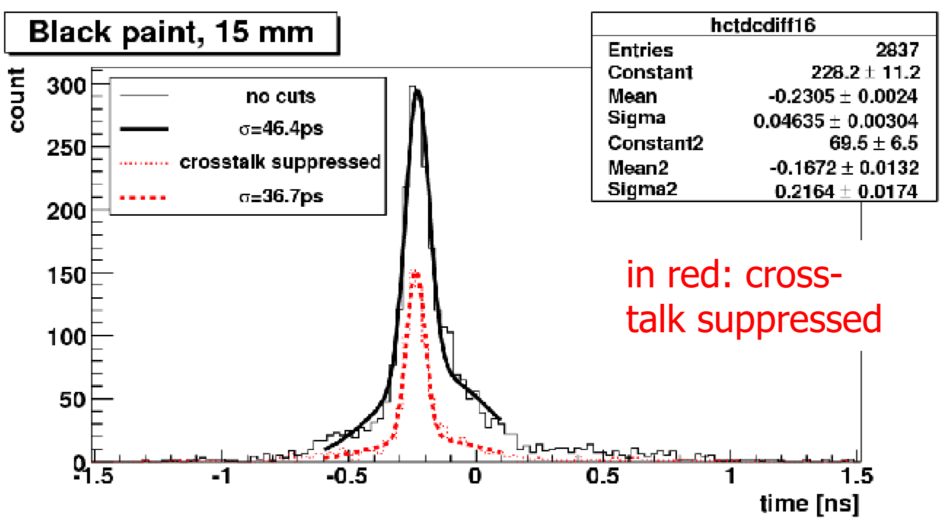
TOF-PET with Cherenkov light detection: proof of principle

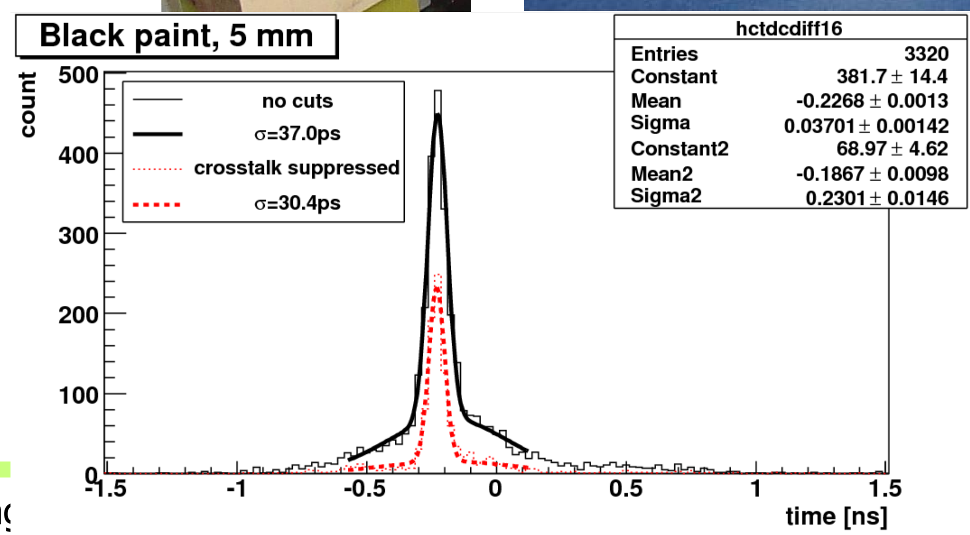
S. Korpar et al, NIM A654 (2011) 532

Two detectors (back-to-back)
- 25 x 25 x 15 mm^3 crystals (black painted or Teflon wrapped)
- MCP-PMT (Hamamatsu, same as in the Belle II TOP counter)
- ^{22}Na source



- 15 mm long crystal: FWHM ~ 95 ps
- 5 mm long crystal: FWHM ~ 70 ps





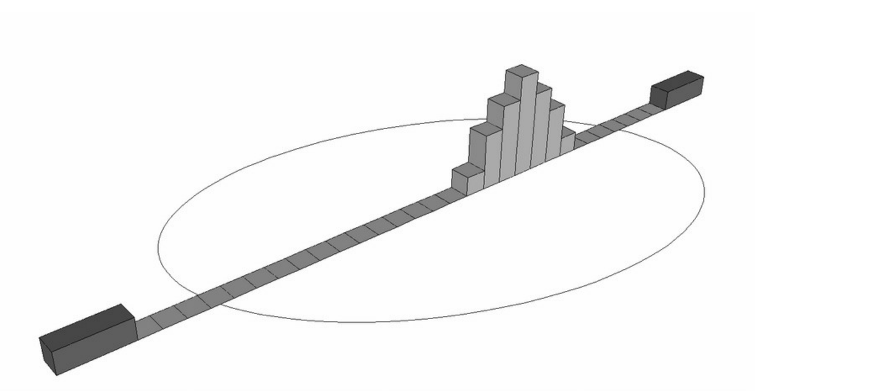
Point source position

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

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

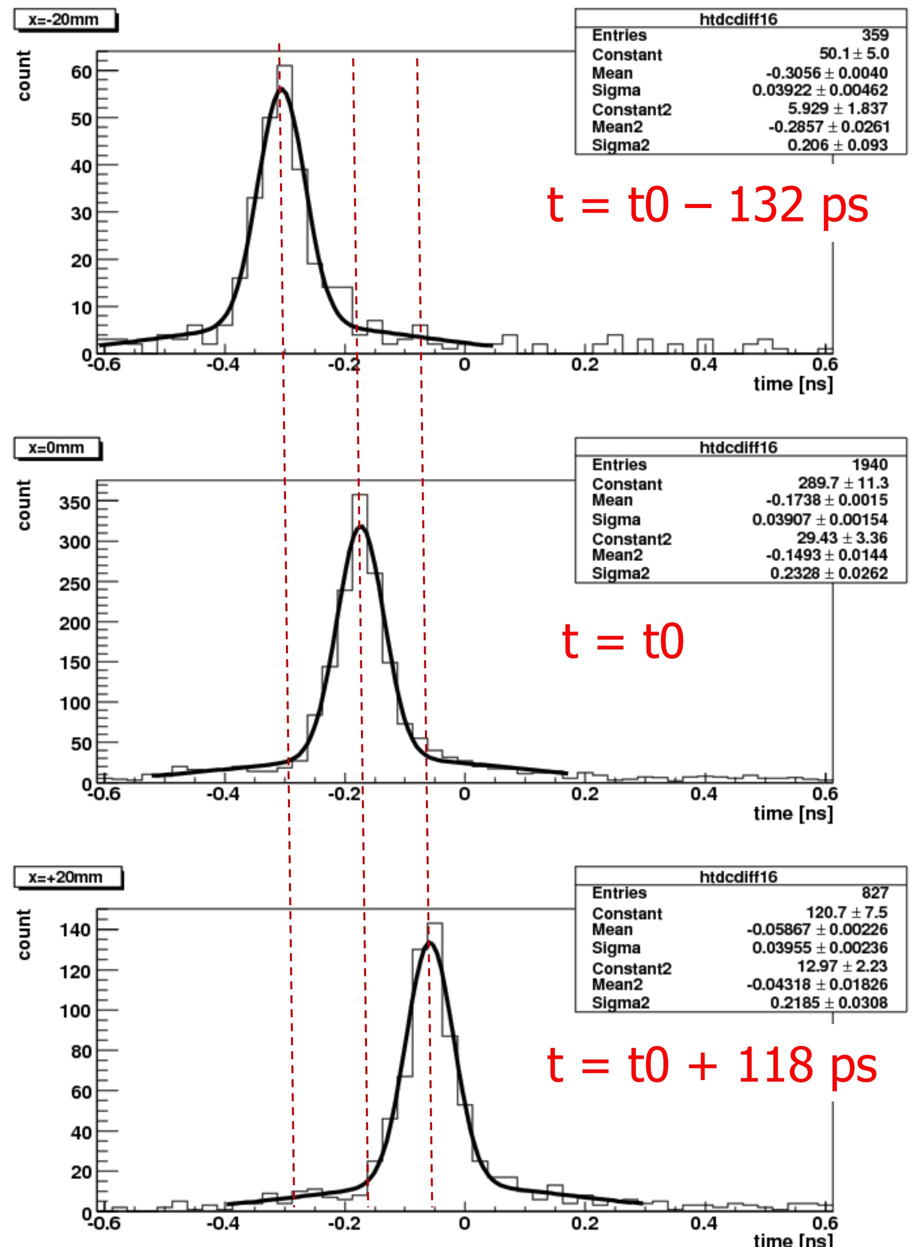
  ~ 95 ps FWHM
- position resolution along the line of response σ ~ 6 mm rms,

  ~ 14 mm FWHM

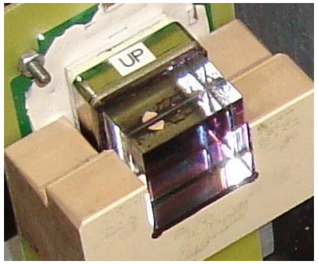


Black painted 15 mm PbF_2 crystals.

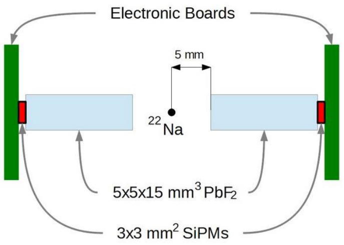
S. Korpar et al, NIM A654 (2011) 532

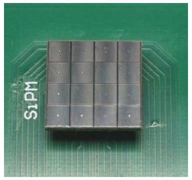


Some results of our studies



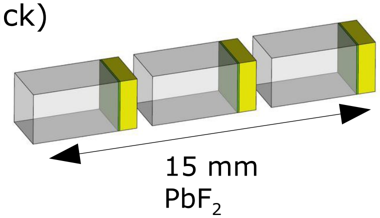
(25 * 25 * **15**) mm³ **PbF_2** (black)
+ (22.5 * 22.5) mm² **MCP-PMT**





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**]



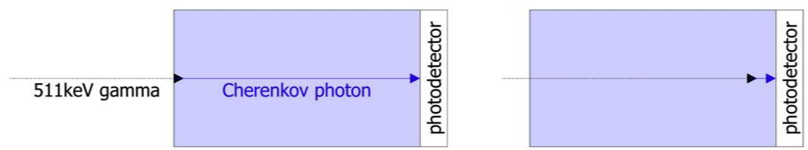
Result	Reference
Cherenkov TOF PET, TOF: **95 ps** FWHM	Korpar, NIM A 654 (2011) 532
With SiPMs, TOF: **306 ps** FWHM	Dolenec, IEEE TNS 63:5 (2016) 2478
Cherenkov PET module: Single side efficiency: **35 %**	Dolenec, NIM A 952 (2020) 162327
Multi-layer detector (simulation) TOF: **22 ps** FWHM before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

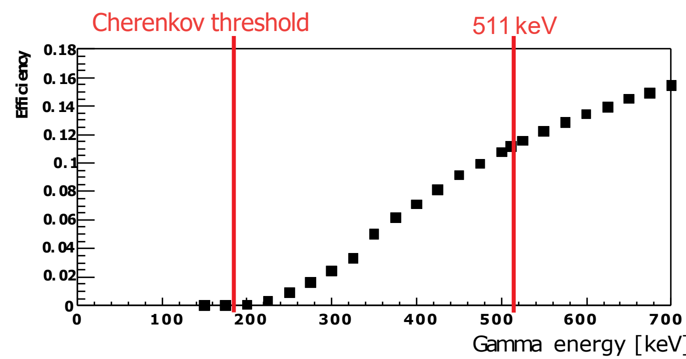
- **Optical photon travel time spread** in the crystal
  - remaining limitation to TOF resolution



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

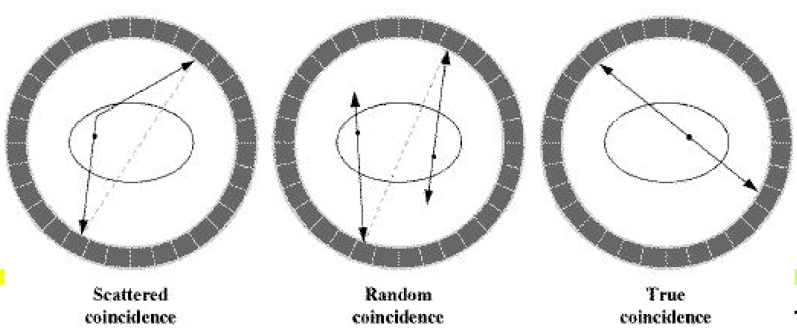
➡ **multi-layer**

- 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?

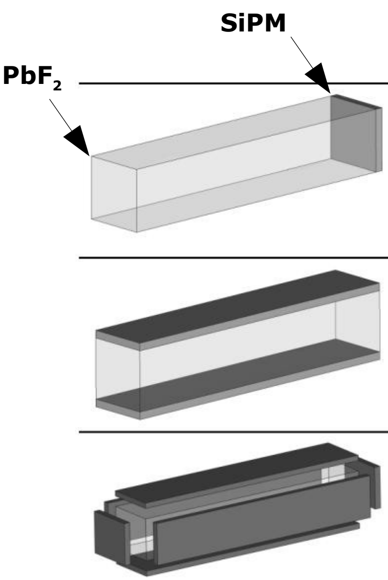


Essential question → MC simulation to evaluate the effect →

Crystal readout configurations

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

Simulation:
- 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
1-sided-back	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
2-sided-top-bottom	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

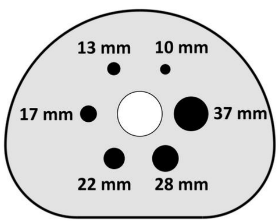
Figure-of-merit: $FOM = \frac{\epsilon^2}{CTR}$

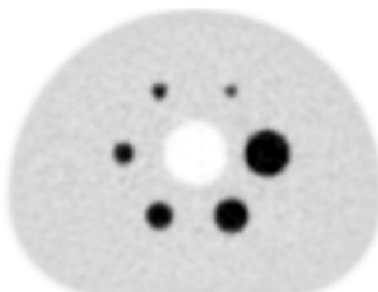
SPTR = single photon time resolution

G. Razdevšek *et al.*, "Exploring the Potential of a Cherenkov TOF PET Scanner: A Simulation Study," IEEE TRPMS 7 (2023) 52, doi: 10.1109/TRPMS.2022.3202138.

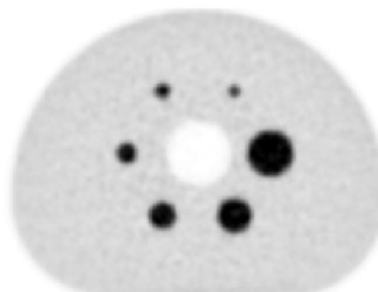
Results: Image Quality

NEMA image quality phantom

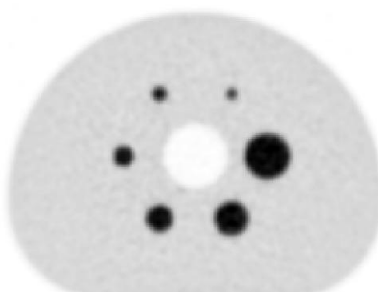




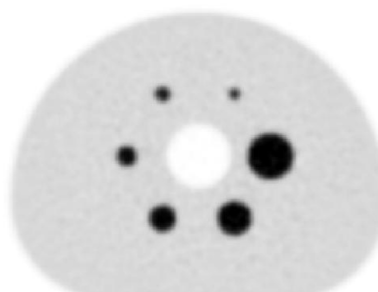
Reference-scanner-214ps



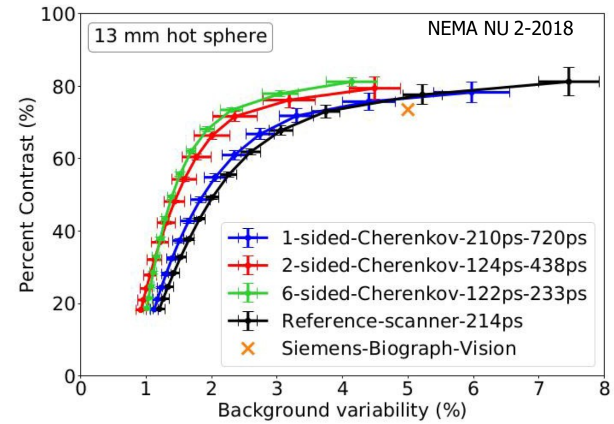
1-sided-210ps-720ps

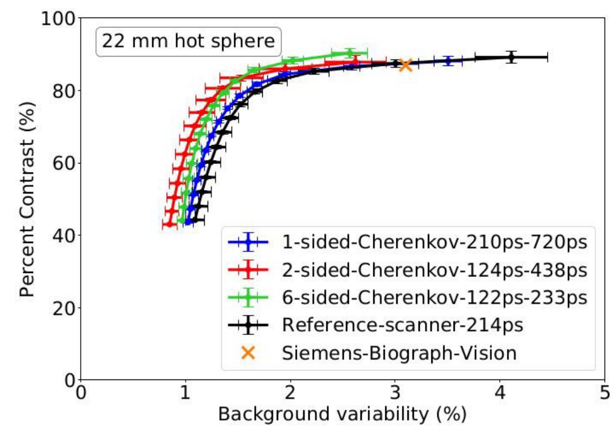


2-sided-124ps-438ps



6-sided-122ps-233ps





Better sensitivity and CTR compensate for higher scatter fraction

Image quality comparable to state-of-the-art

G. Razdevšek *et al.*, IEEE TRPMS 7 (2023) 52

Hybrid Cherenkov-scintillator approach

Cherenkov light is also produced in scintillator materials

- BGO: low cost, high density, historically used for PET

The idea:

- use (abundant) scintillation light for energy measurement/photopeak cut
- use the few Cherenkov photons for timing

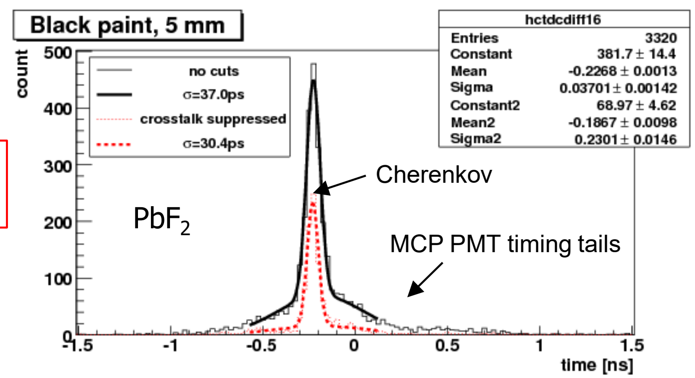
Issues:

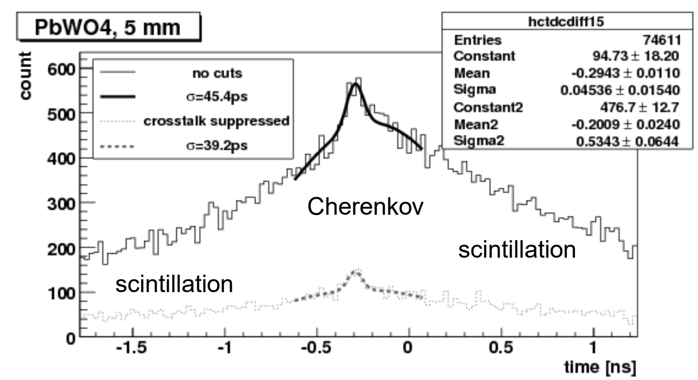
- few coincidences formed by detection of Cherenkov photons on both sides
- → long tails in CTR distribution, how to handle them in reconstruction?

→ does not look very promising

Measured CTR with PbF_2 vs. $PbWO4$ scintillator:

S. Korpar, NIM A 654 (2011) 532

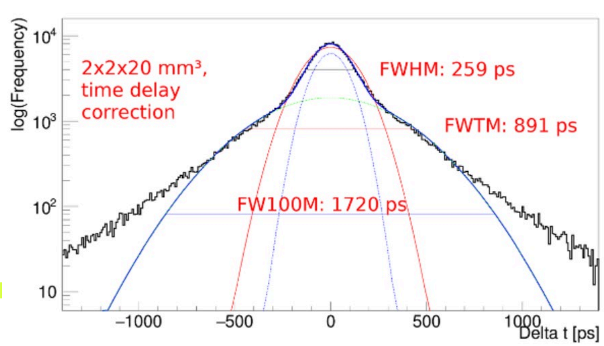


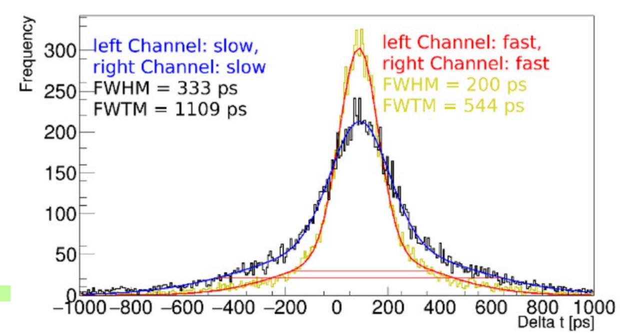


Measured CTR with BGO:

N. Kratochwil et al., Phys. Med. Biol. 65 (2020) 115004

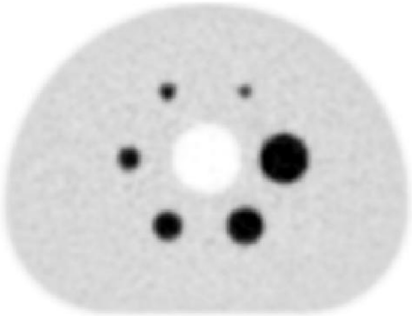
N. Kratochwil et al., IEEE TRPMS 5(5) (2020) 619-629



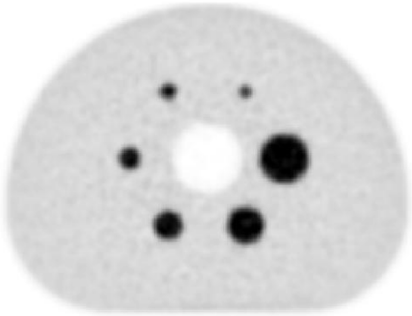


Cherenkov based scanners, conclusion

- 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 R. Ota, Phys. Med. Biol. 64 (2019) 07LT01
  - detection efficiency (module) of 35% R. Dolenec et al, NIM A 952 (2020) 162327
- Cherenkov TOF-PET scanner simulations
  - better sensitivity and CTR compensate for higher scatter fraction
  - **image quality comparable to state-of-the-art**
- Advanced detector geometries (2-sided top-bottom, multi-layer)
  - even better image quality



Reference-scanner-214ps

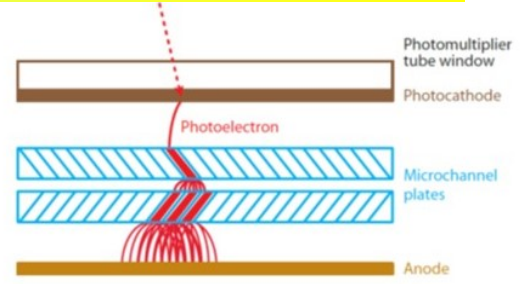


1-sided-210ps-720ps

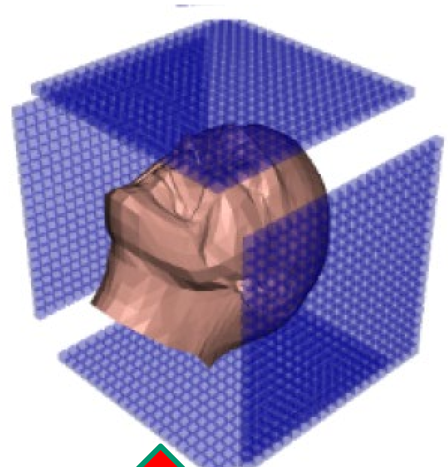
G. Razdevšek *et al.*, IEEE TRPMS 7 (2023) 52

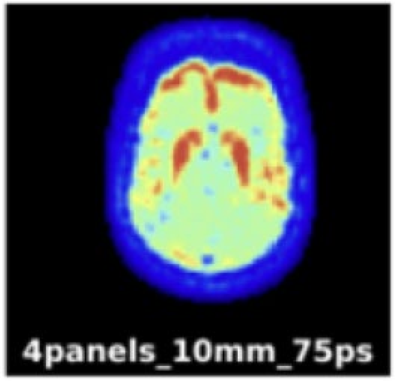
Cherenkov-based TOF-PET with a large area MCP-PMT

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



LAPPD with PbF_2 crystals





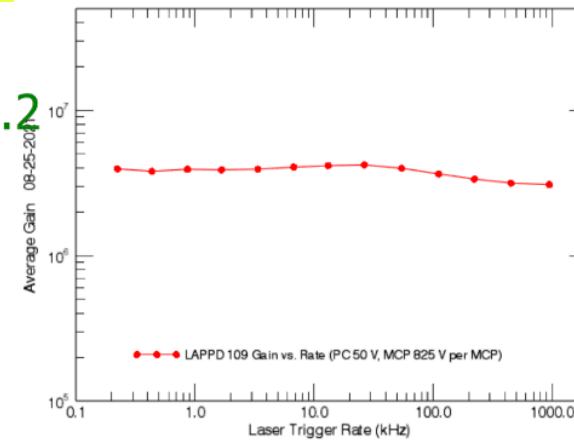
CherPET: an ERC (European Research Council) Proof-of-Principle project

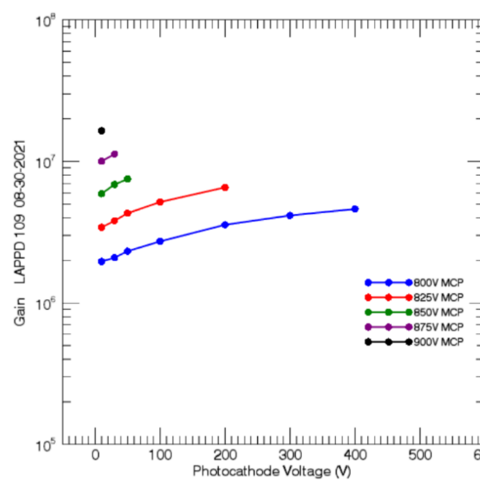
LAPPD with PbF_2 crystals attached to the entry window: an almost ideal flat panel device

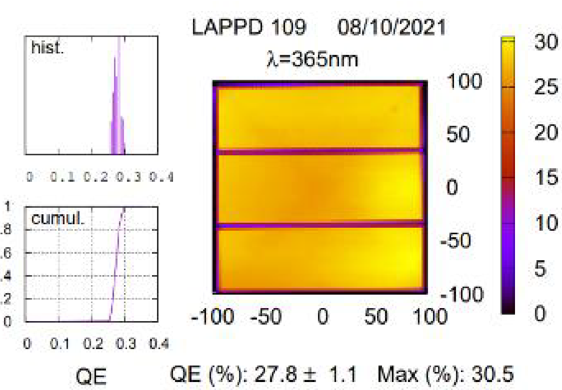
LAPPD (large area picosecond photodetector) Gen II

Characteristics (Incom):

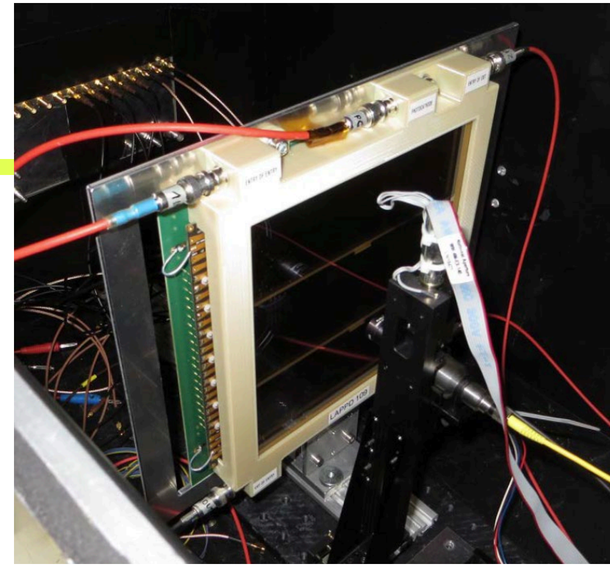
- size 230 mm x 220 mm x 22 mm (243 mm x 274 mm x 25.2 mm with mounting case)
- borosilicate back plate with interior resistive ground plane anode – 5 mm thick
- capacitively coupled readout electrode
- MCPs with 20 μm pores at 20 μm pitch
- two parallel spacers (active fraction ≈ 97 %)
- gain $\approx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





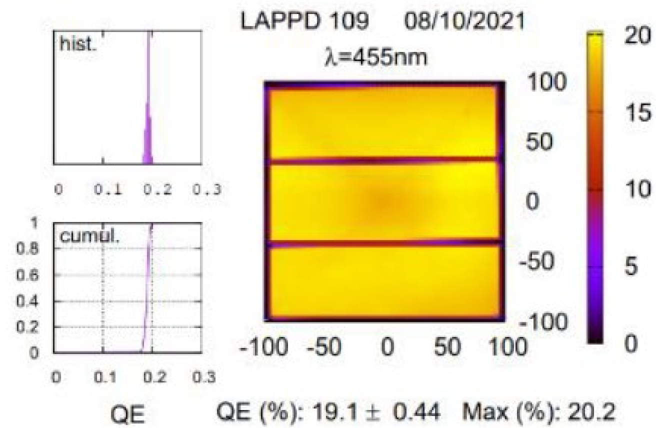


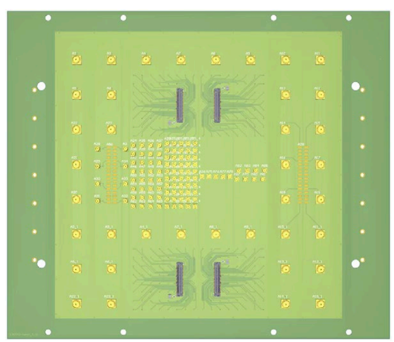
LAPPD evaluation

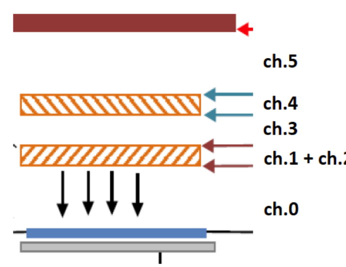


- Two 10 μm devices acquired
- LAPPD installed in the dark box:
- CAEN HiVolta (DT1415ET), 8 Ch Reversible 1 kV/ 1 mA Desktop HV Power Supply – floating channels
- Standard setup with QDC, TDC, 3D stage ...
  - TDC value corrected for time-walk
- ALPHALAS PICOPOWER™-LD Series of Picosecond Diode Lasers – 405 nm
  - FWHM ≈20ps
  - light spot diameter on the order of 100μm
- Measure response in the lab with modular electronics, FastIC and PETSys
- Custom segmentation

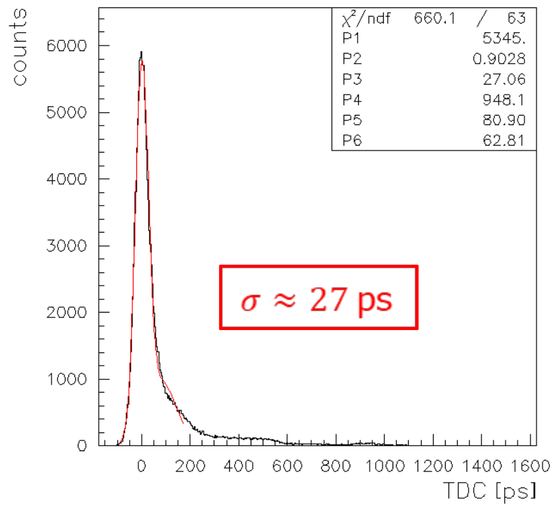
to study the capacitive coupling and the charge spread







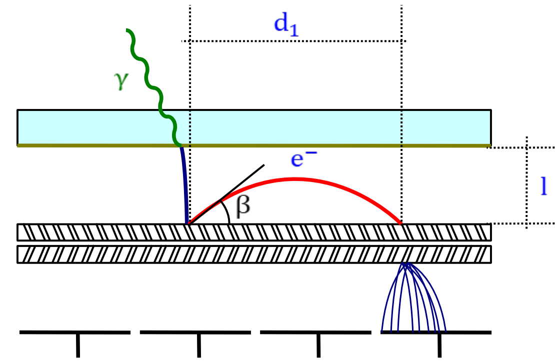
Characterizing LAPPD: time resolution

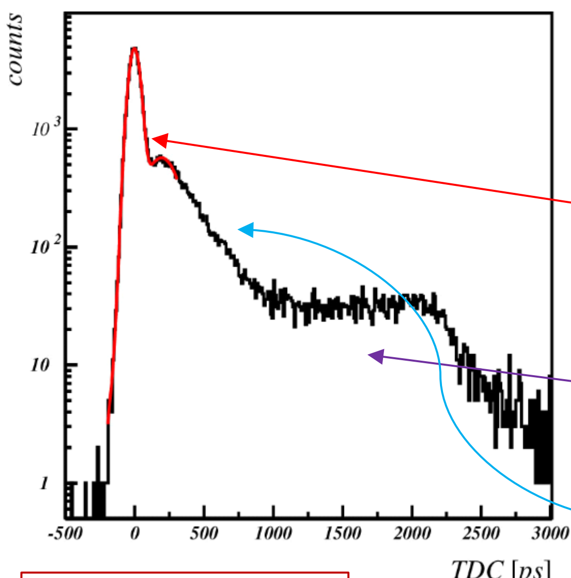


Photoelectron back-scattering produces a rather long tail in timing distribution and position resolution.

Photoelectron backscattering reduces collection efficiency and gain, and contributes to cross-talk in multi-anode PMTs

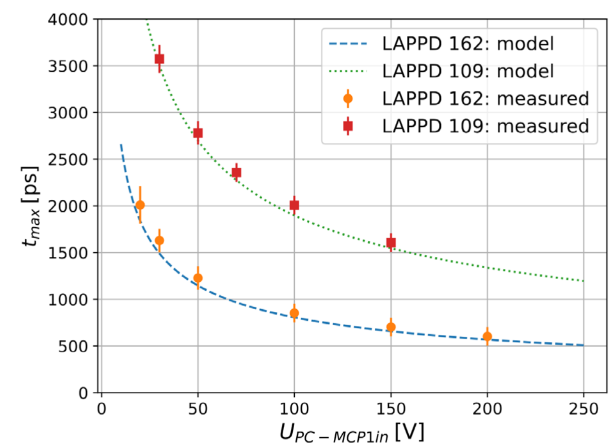
R. Dolenec *et al.*, NIM A 1069 (2024) 169864





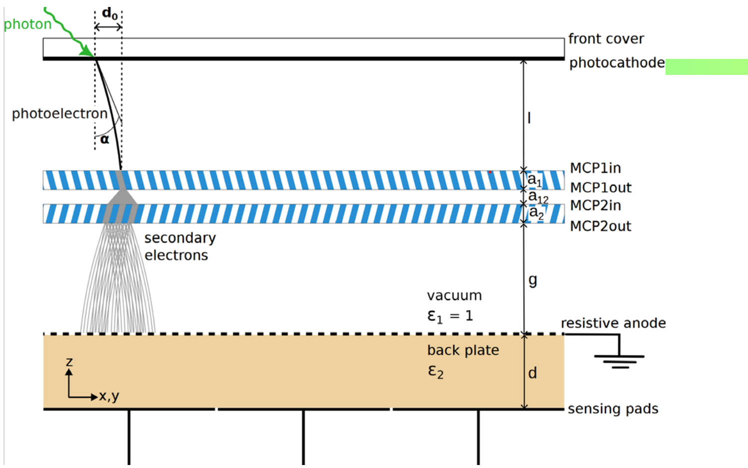
Typical single photon timing distribution with a narrow main peak (σ~30-40 ps) and contributions from photoelectron elastic back-scattering (flat distribution) and inelastic back-scattering.

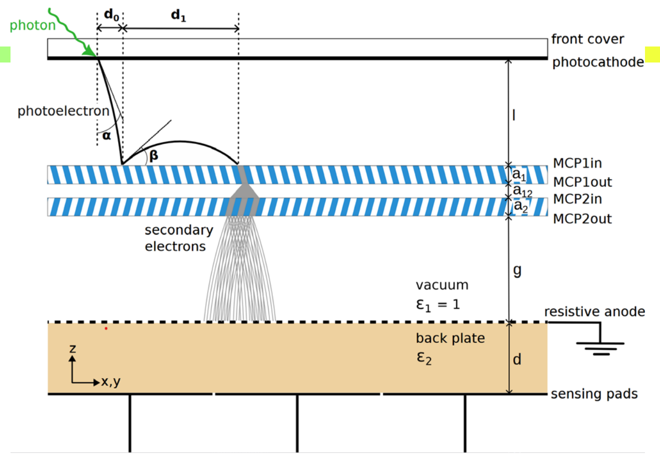
S.Korpar et al, PD07

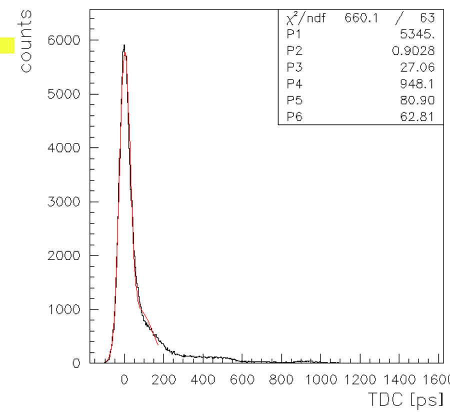


S.Korpar et al, arXiv:2406.19421[physics.ins-det] submitted to NIMA

Characterizing LAPPD: time resolution and charge sharing

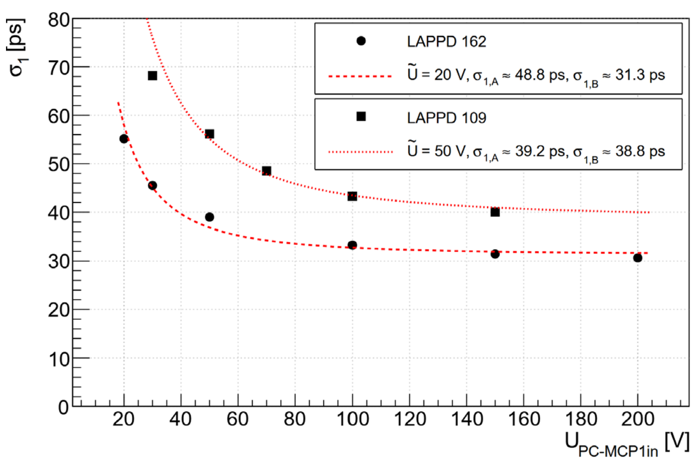




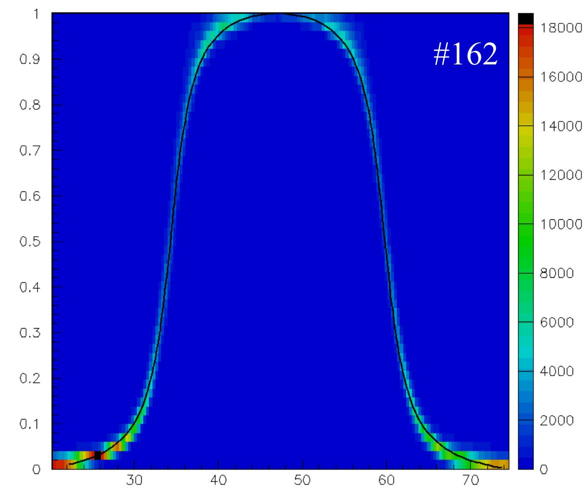


σ(ps)

Time resolution vs PC-MCP1 voltage *U*



S.Korpar et al, arXiv:2406.19421[physics.ins-det] submitted to NIMA



Charge sharing – capacitively coupled read-out electrodes. Black: model prediction,using Shockley-Ramo theorem.

Next steps: finalize read-out, attach crystals, test the back-to-back configuration

Summary

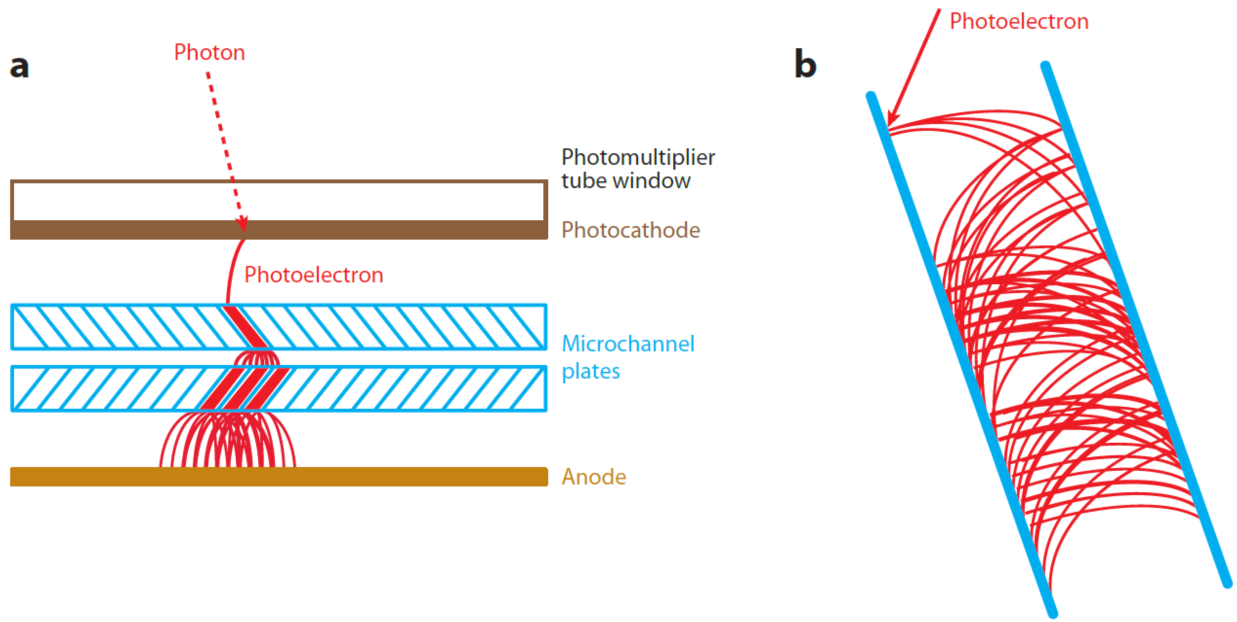
The interplay of detector R&D for particle physics and medical imaging has a long history, and this will remain one of the sources of innovation in medical imaging.

Limited angle devices with very fast gamma detection look very promising – 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.

More slides

Micro Channel Plate PMT (MCP-PMT)



~10-100 μm diameter

Multiplication step: a continuous dynode – a micro-channel coated with a secondary emitter material

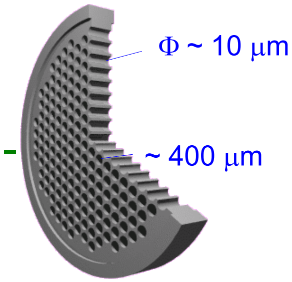
Micro Channel Plate PMT (MCP-PMT)

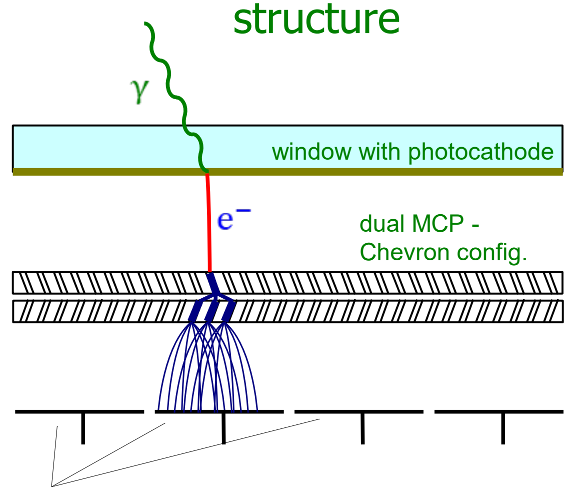
Similar to ordinary PMT – the dynode structure is replaced by MCPs.

Basic characteristics:

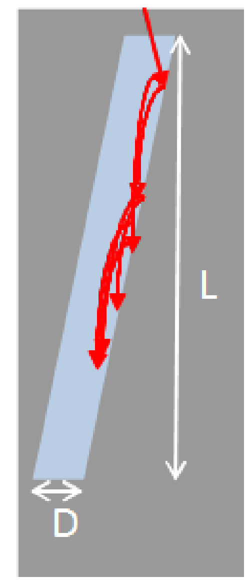
- Gain ~ 10^6 → single photon
- Collection efficiency ~ 60%
- Small thickness, high field → small TTS
- Works in magnetic field
- Segmented anode
  → position sensitive

MCP is a thin glass plate with an array of holes (<10-100 μm diameter) - - a continuous dynode structure

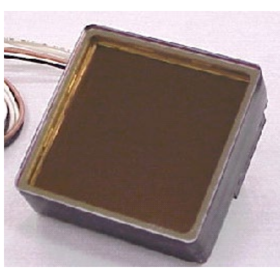




Anodes → can be segmented according to application needs



MCP gain depends on L/D ratio – typically 1000 for L/D=40

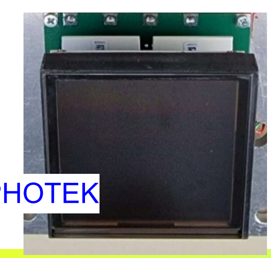


PHOTONIS

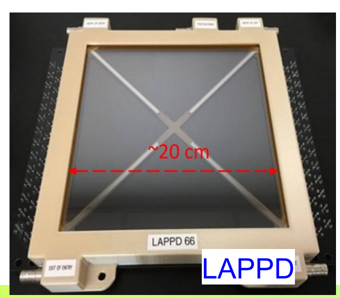
HAMAMATSU

1□ □

2□ □



PHOTEK

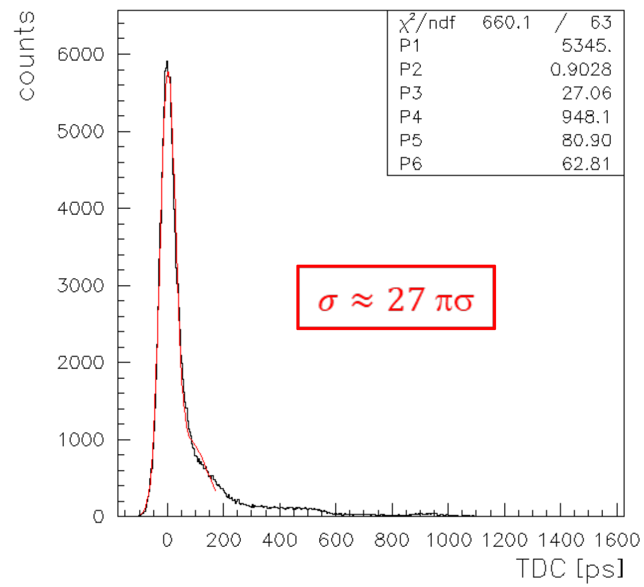


LAPPD

8□ □

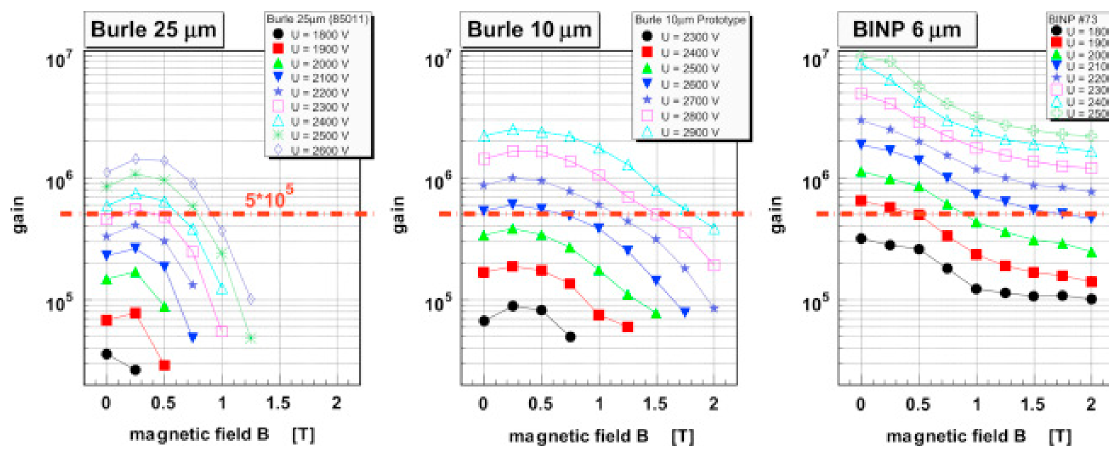
Micro Channel Plate PMT: properties

Very fast: single photon detection with sigma of ~30-40 ps

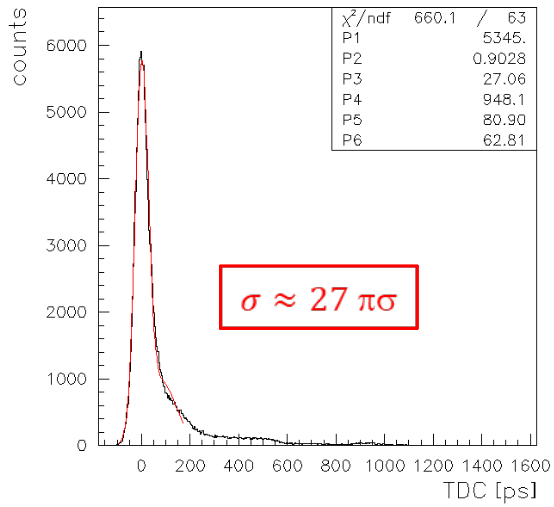


MCP PMTs work well in magnetic fields

→ performance depends on the diameter of the micro-channels

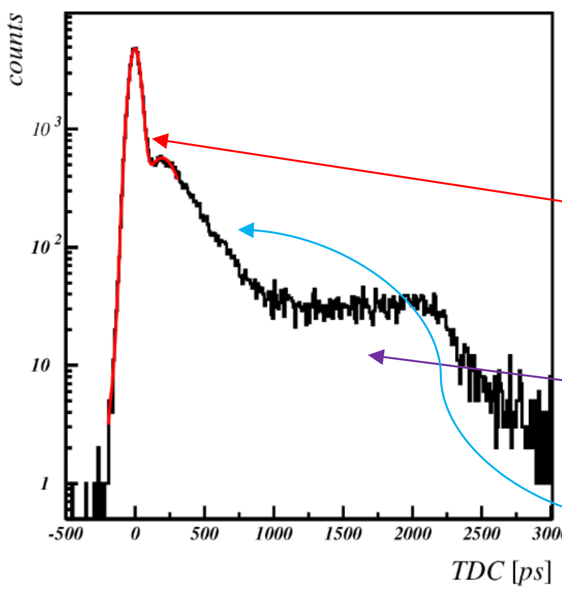


MCP-PMT: single photon pulse height and timing

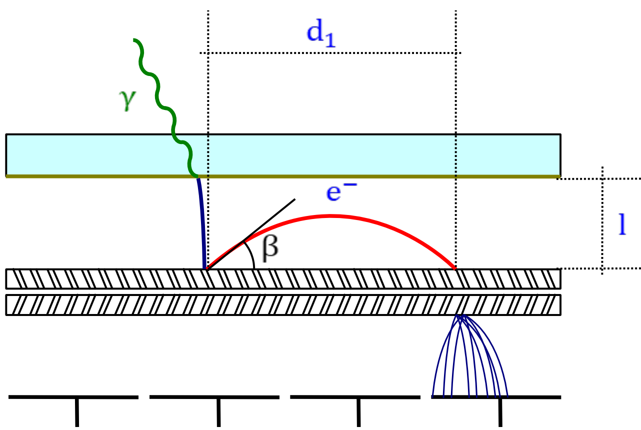


Photoelectron back-scattering produces a rather long tail in timing distribution and position resolution.

Photoelectron backscattering reduces collection efficiency and gain, and contributes to cross-talk in multi-anode PMTs



Typical single photon timing distribution with a narrow main peak (σ~30-40 ps) and contributions from photoelectron elastic back-scattering (flat distribution) and inelastic back-scattering.



S.Korpar@PD07

Modelling MCP-PMT: Photoelectrons in a uniform electric field

Photoelectrons travel from the photocathode to the electron multiplier (uniform electric field $\frac{U}{l}$, initial energy $E_0 \ll Ue_0$):

- photoelectron range

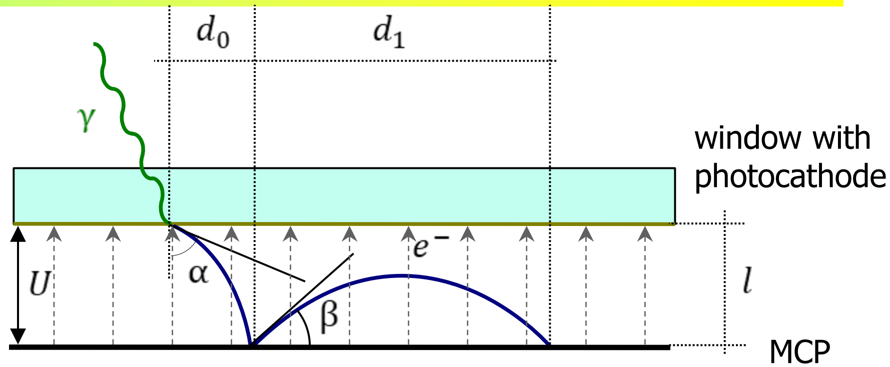
$$d_0 \approx 2l\sqrt{\frac{E_0}{Ue_0}}sin(\alpha)$$

- and maximal travel time (sideway start)

$$t_0 \approx l\sqrt{\frac{2m_e}{Ue_0}}$$

- time difference between downward and sideways initial direction

$$\Delta t \approx t_0\sqrt{\frac{E_0}{Ue_0}}$$



Backscattering delay and range (maximum for elastic scattering):

- maximum range vs. angle

$$d_1 = 2lsin(2\beta)$$

  maximum range for backscattered photoelectron is twice the photocathode – first electrode distance
- maximum delay vs. angle

$$t_1 = 2t_0sin(\beta)$$

  maximum delay is twice the photoelectron travel time
- time of arrival of elestically scattered photoelectrons: flat distribution up to max $t_1 = 2t_0$

Example ($U = 200$ V, $E_0 = 1$eV, $l = 6$mm)

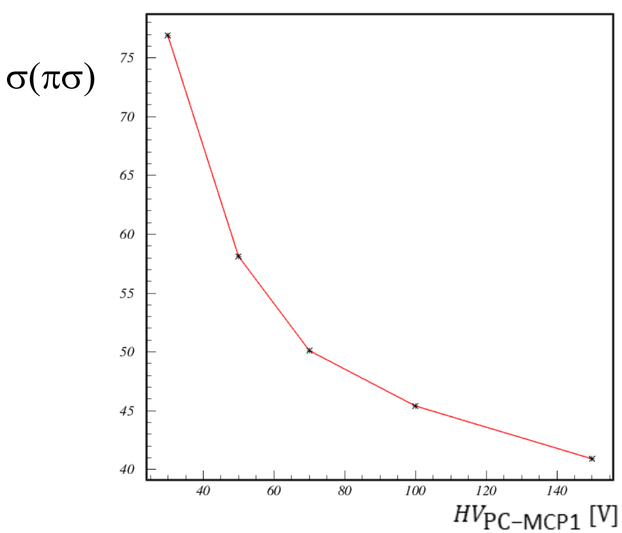
photoelectron:

- max range $d_0 \approx 0.8$ mm
- p.e. transit time $t_0 \approx 1.4$ ns
- $\Delta t \approx 100$ ps

backscattering:

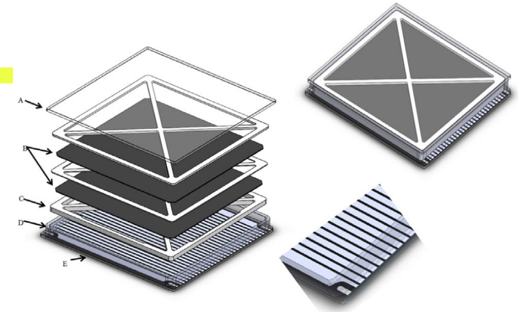
- max range $d_1 = 2l = 12$ mm
- max delay $t_1 = 2.8$ ns

S.Korpar@PD07

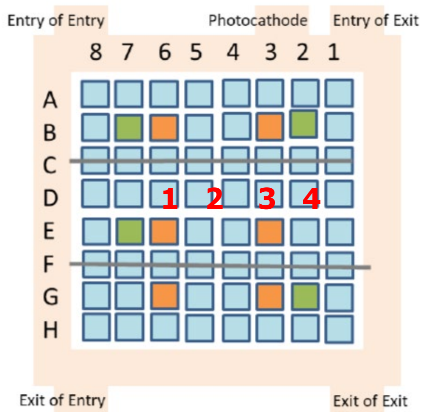


Time resolution vs PC-MCP1 voltage

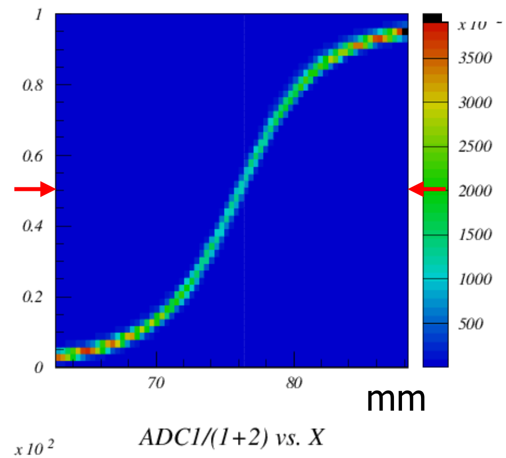
LAPPD – charge sharing in Gen II capacitively coupled electrode readout

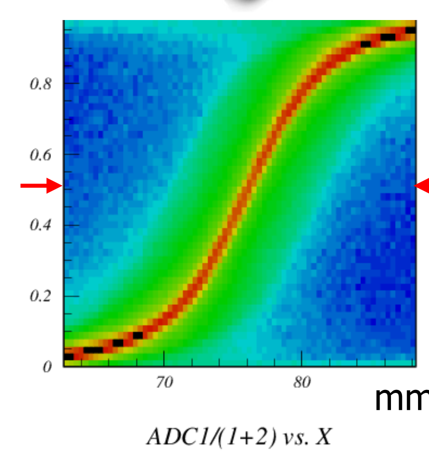


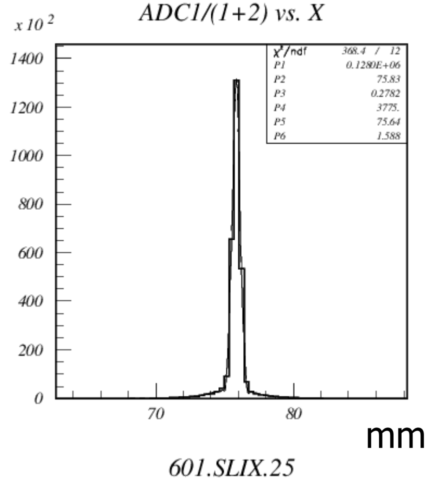
- fraction of the signal on channel 3 vs laser spot x position: $f(x) = \frac{q_3}{\sum_i q_i}$
- scan between the centres of pads 2 and 3 (top)

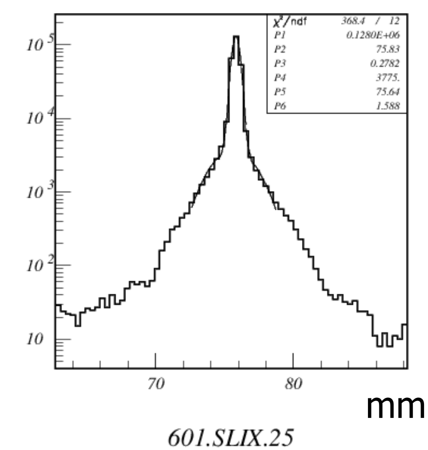


- central slice where signal is equally split between the pads (bottom)
- narrow peak is due to the light spot size and photoelectron spread
- longer tail from photoelectron backscattering - ≈ 6 mm on each side $\rightarrow \approx 3$ mm PC – MCP1 distance







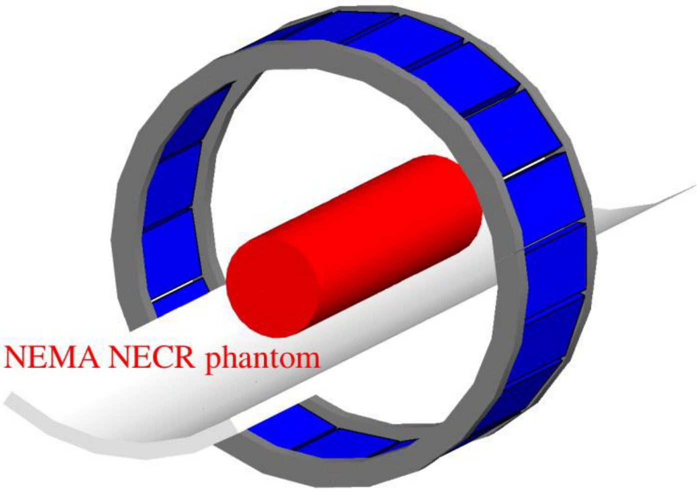


Whole-body scanner simulations

Simulation: GATE v8.1

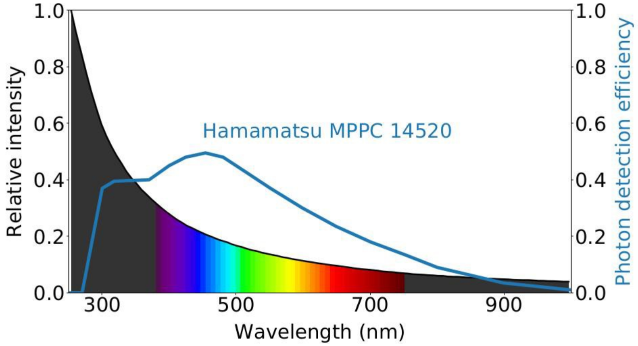
Geometry:

- 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

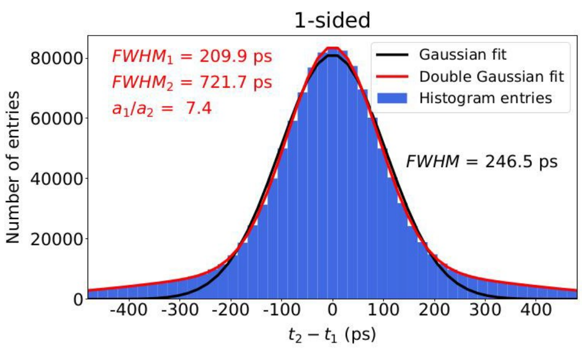


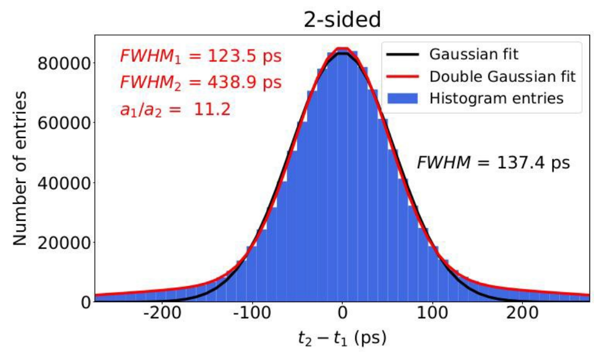
Reconstruction: CASToR v3.1.1

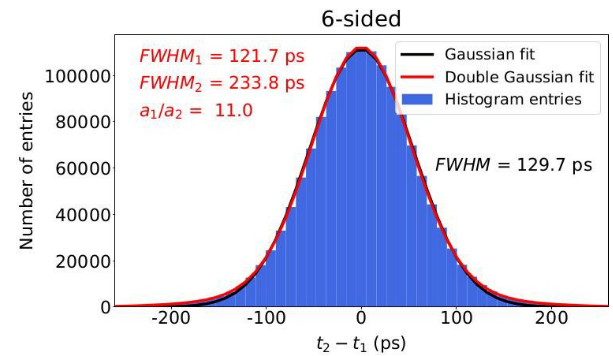
Custom double Gaussian TOF kernel

OSEM-8it:5sub, 1.6 mm voxel, 5 mm filter

Results: CTR distributions



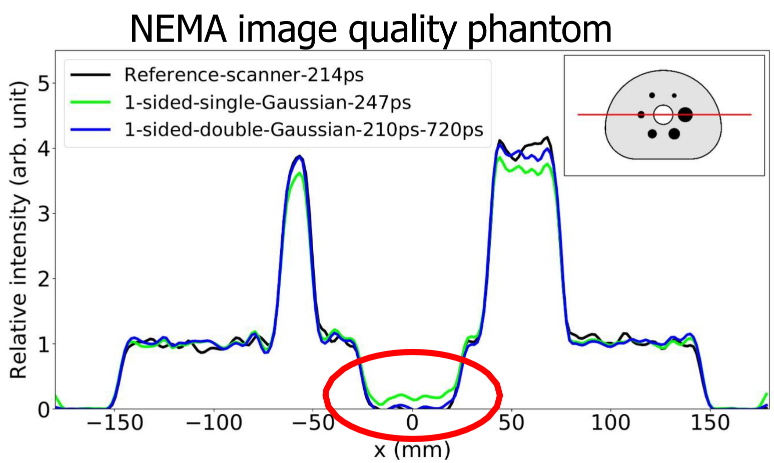




TOF kernel:

-single Gaussian

-double Gaussian



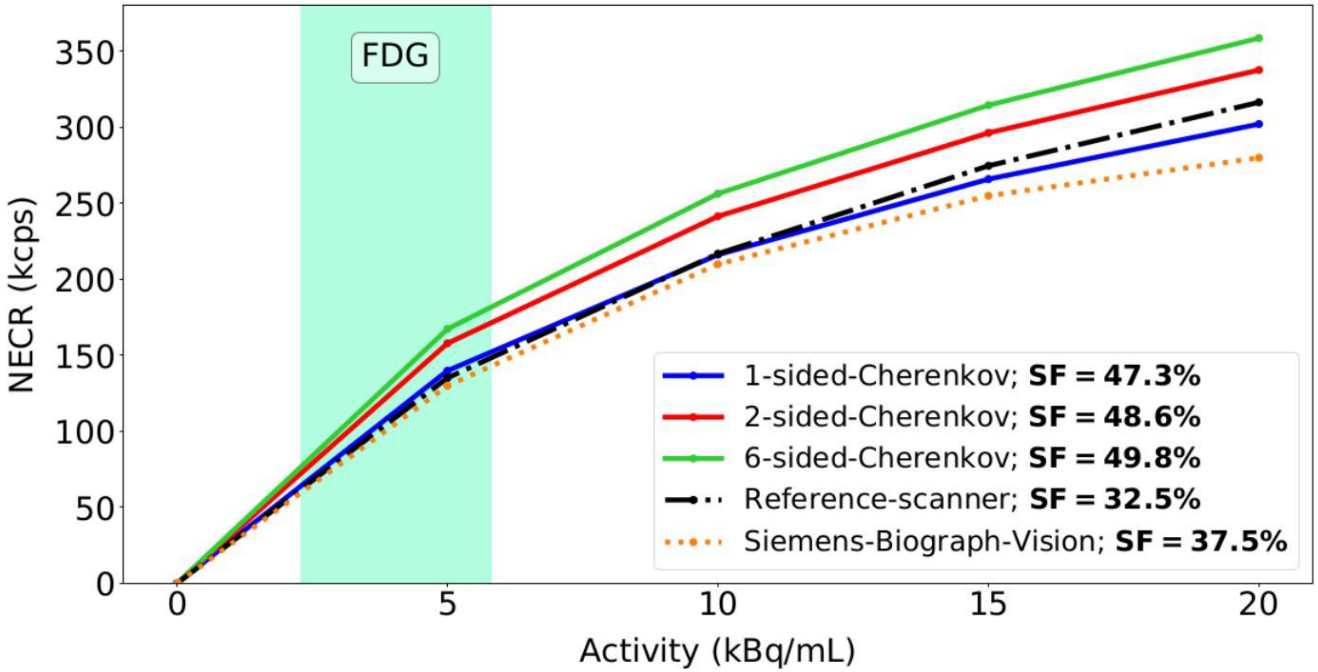
G. Razdevšek *et al.*, IEEE TRPMS 7 (2023) 52

Results: NECR - Noise Equivalent Count Rate

G. Razdevšek *et al.*, IEEE TRPMS 7 (2023) 52

- 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.

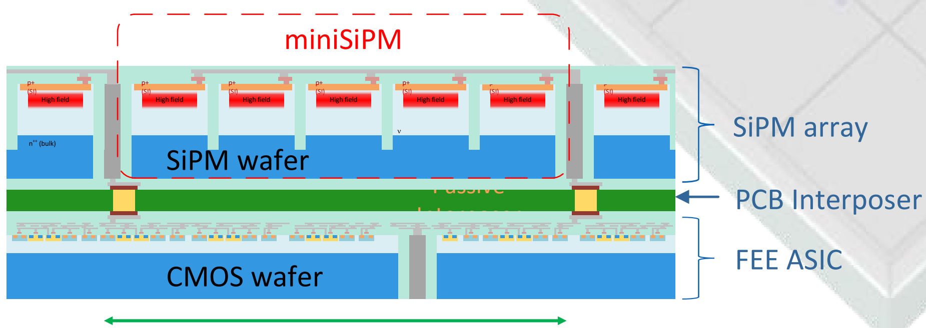
FBK SiPM sensor

2.5D integrated SiPM tile for improved timing

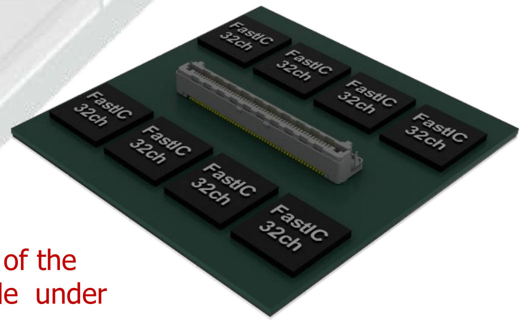
In the short and medium term - medium density interconnection

- 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.

Integrated Photon Detection Module



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



Conceptual drawing of the photon detector module under development