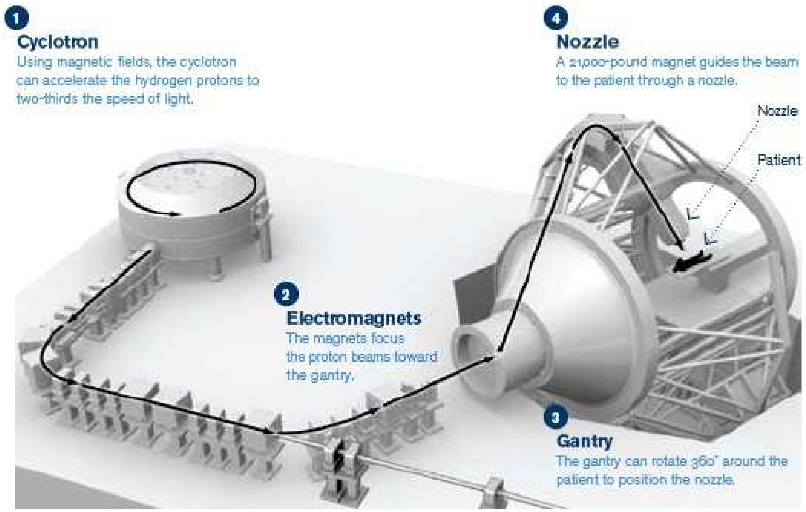
In-vivo dose verification for particle therapy

D.R. Schaart, NCS Lustrum, 5-Oct-2012

medical delta
Erasmus MC
Universitair Medisch Centrum Rotterdam
LUMC
Leids Universitair Medisch Centrum
TU Delft

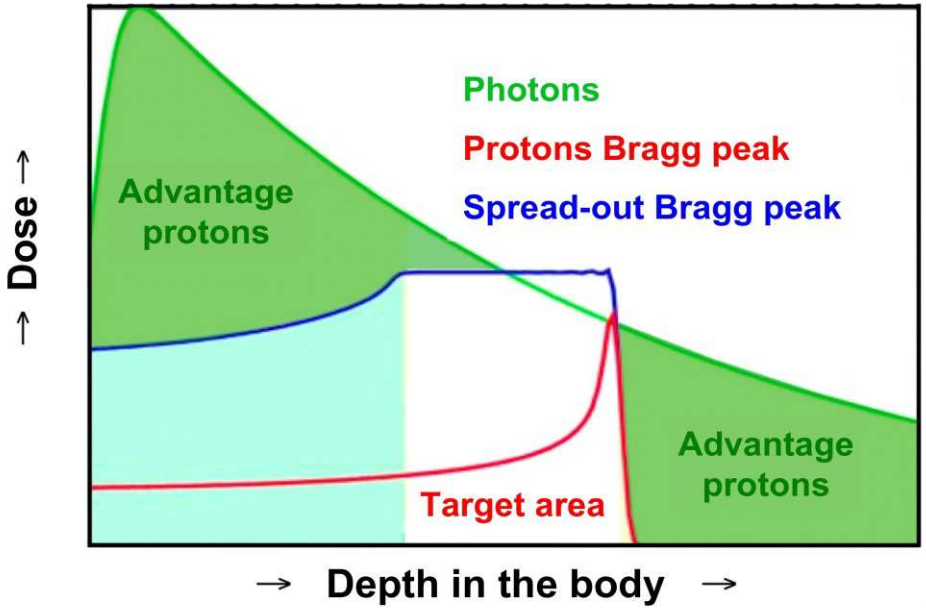
Protons vs. photons

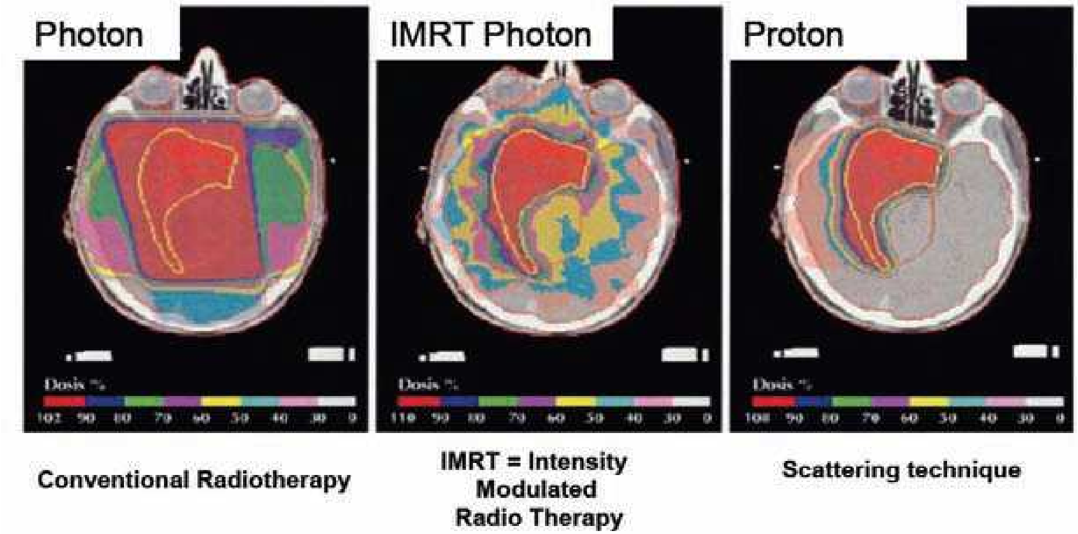


Photons

Protons

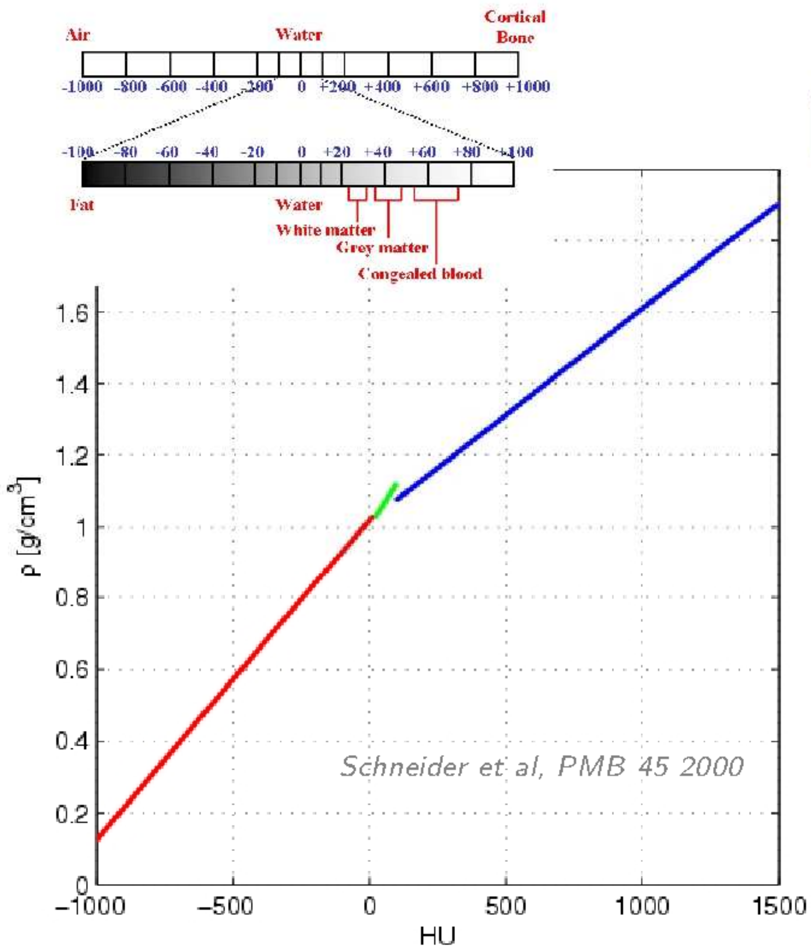
Protons: the promise…





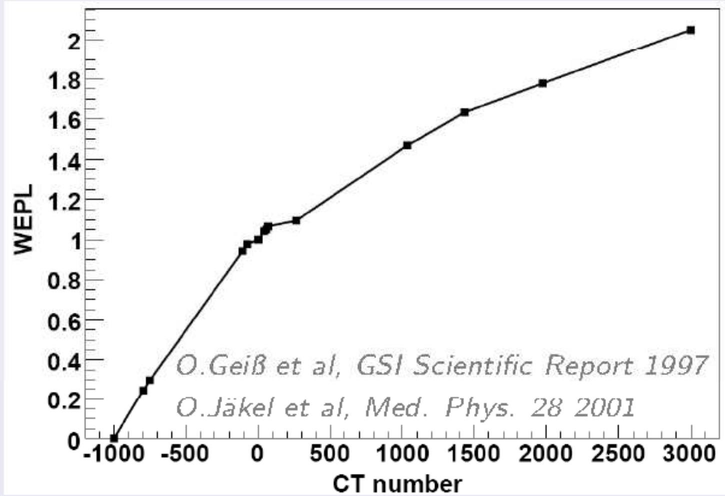
Highly localized dose deposition (Bragg peak) in principle enables more precise dose delivery than photons

... and the problem



$$\rho = \begin{cases} (1.018+0.893\cdot 10^{-3}HU) & HU \leq 14 \\ (1.003+1.169\cdot 10^{-3}HU) & 23 \leq HU \leq 100 \\ (1.017+0.592\cdot 10^{-3}HU) & HU > 100 \end{cases}$$

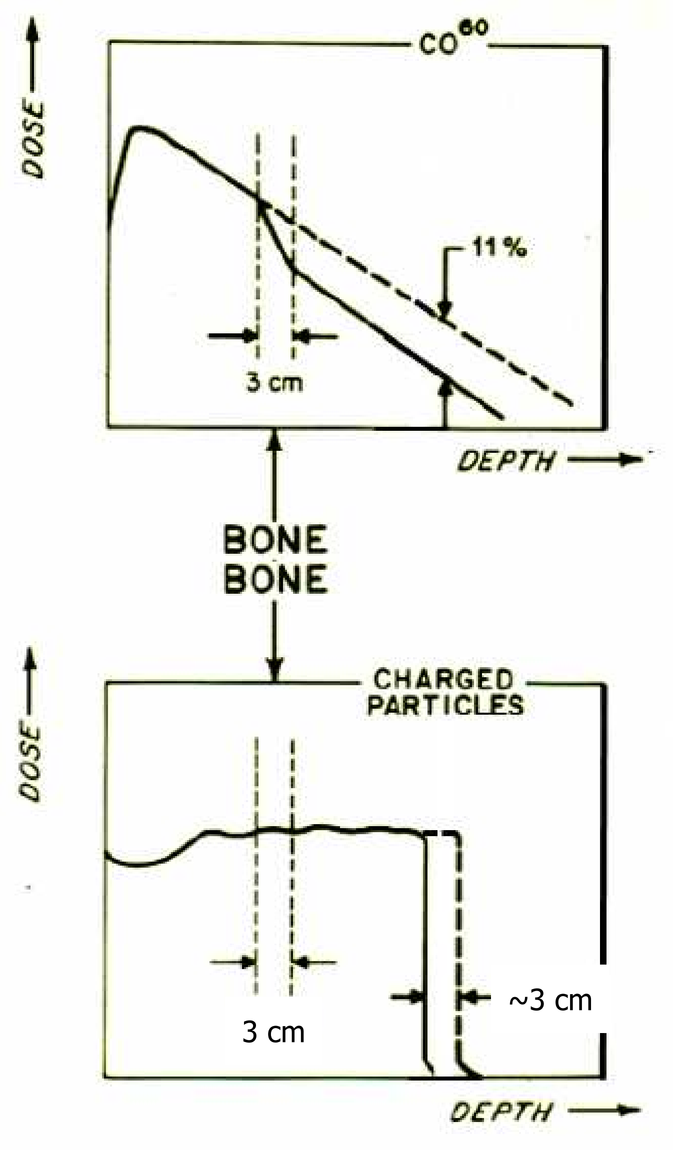
Water Equivalent Path Length



$$if\ I' \cong I_{H_2O} \quad \Rightarrow$$

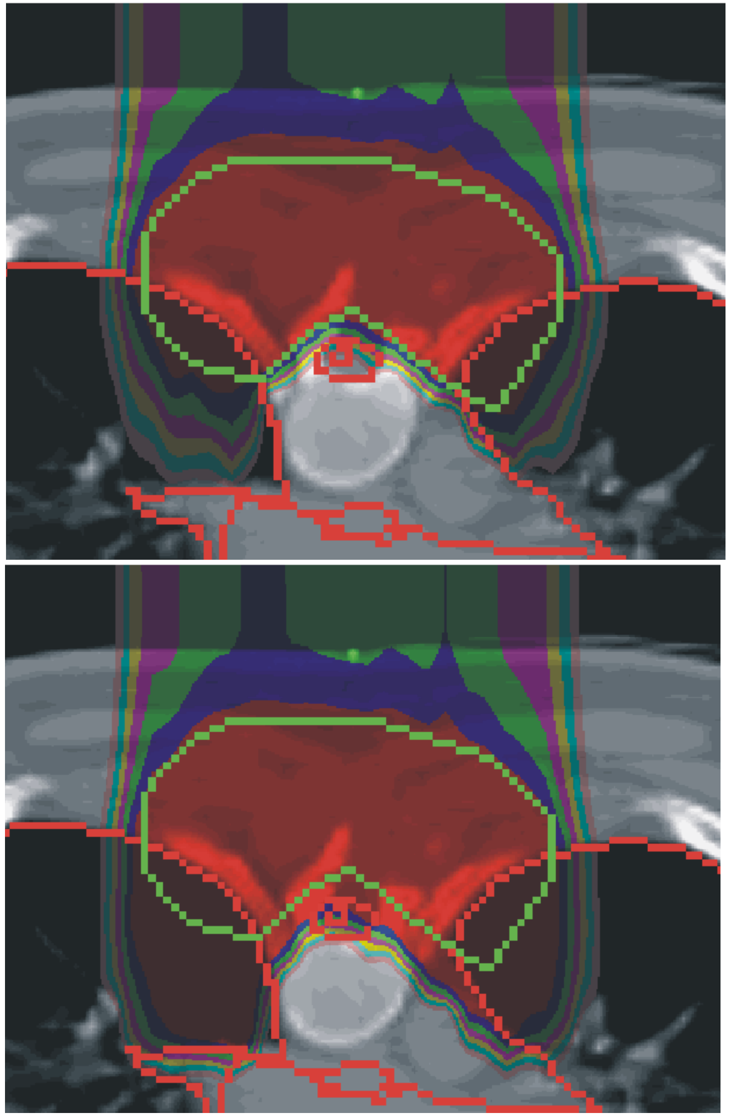
$$z_{wepl} = \int_0^z \frac{\rho(z')}{\rho_{H_2O}}\, dz'$$

Effect of density variations

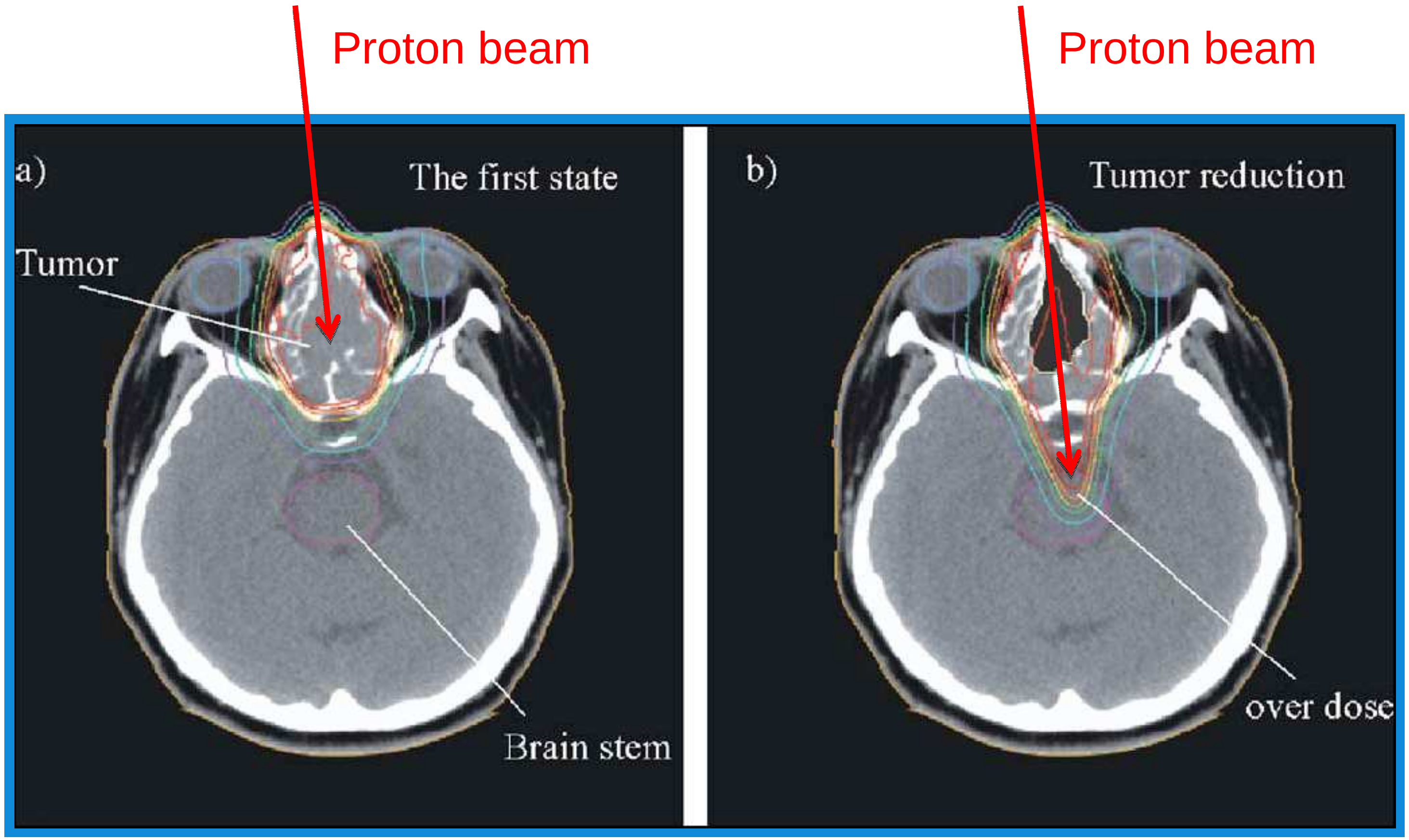


X-rays:
Dose drops
11%

protons:
Range
~3 cm less



Effect of anatomical variations

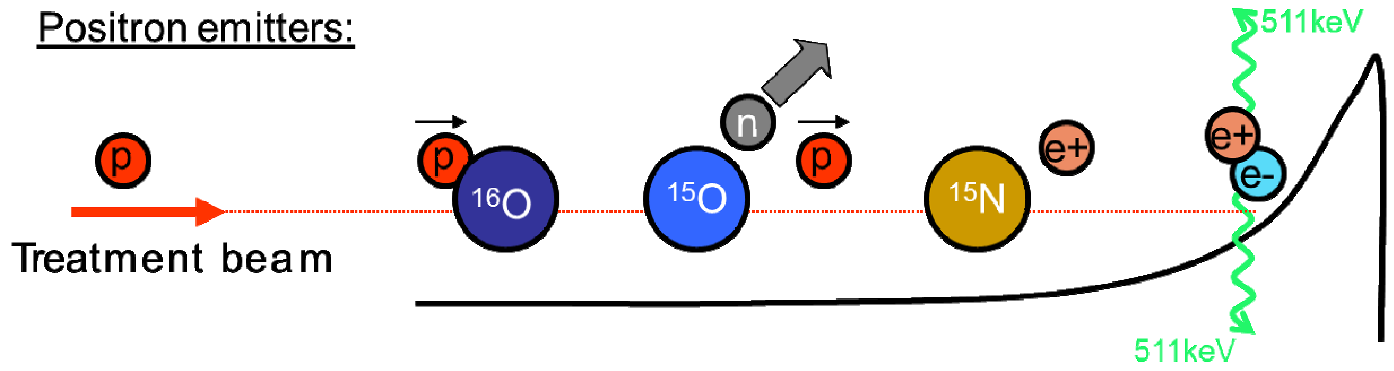


Proton dose distribution for a tumor of the paranasal sinuses. Left: dose distribution at start of treatment. Right: after several fractions. Tumor volume reduction gives rise to overdosing of the brainstem.

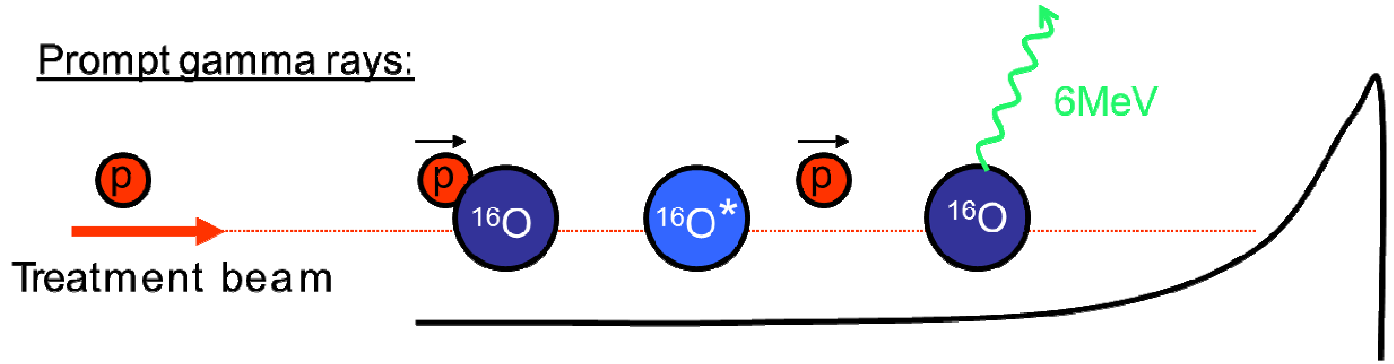
In-situ dose imaging

Human body > 90% Oxygen, Carbon, Hydrogen and Nitrogen

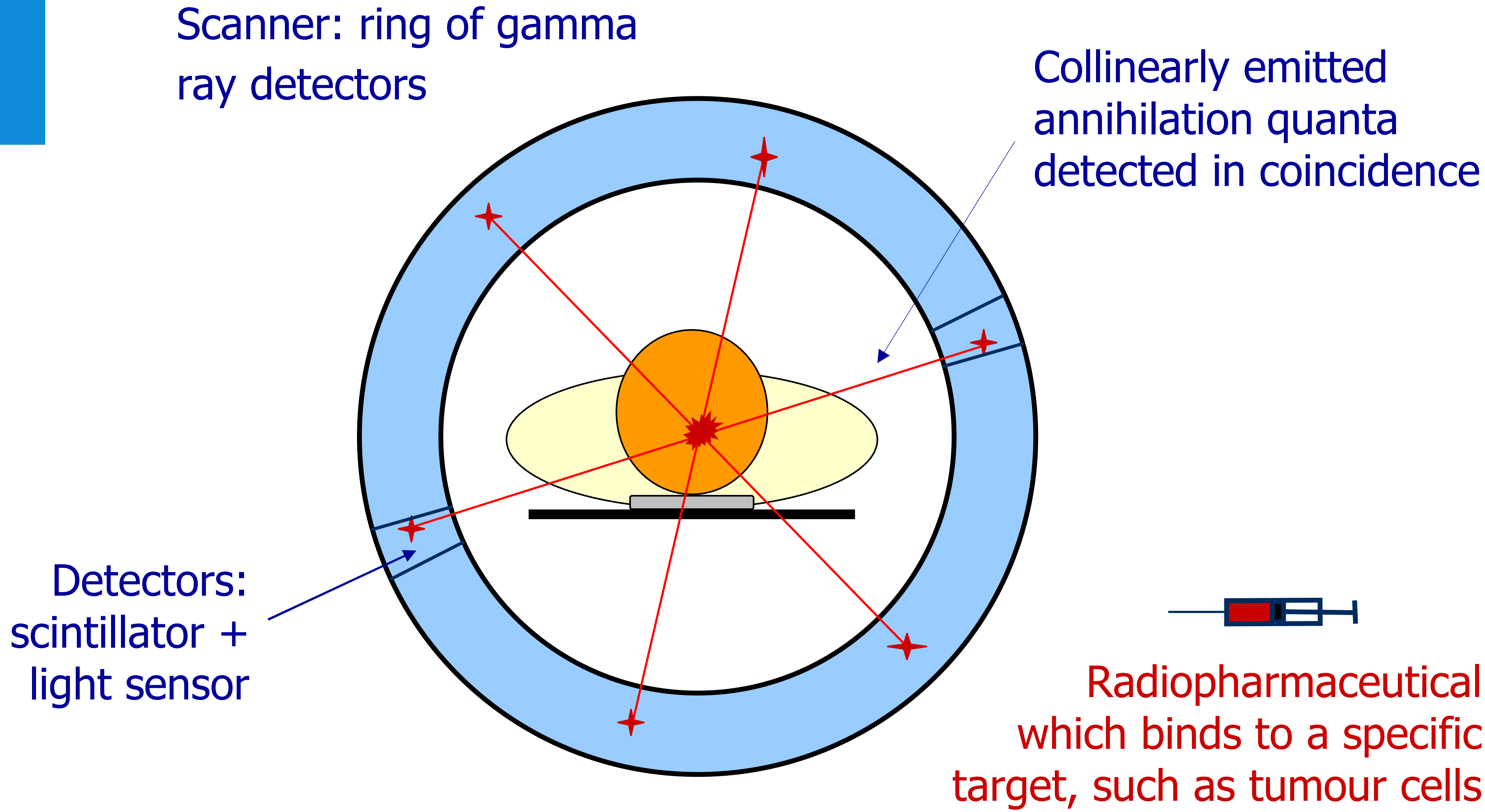
Positron emitters:



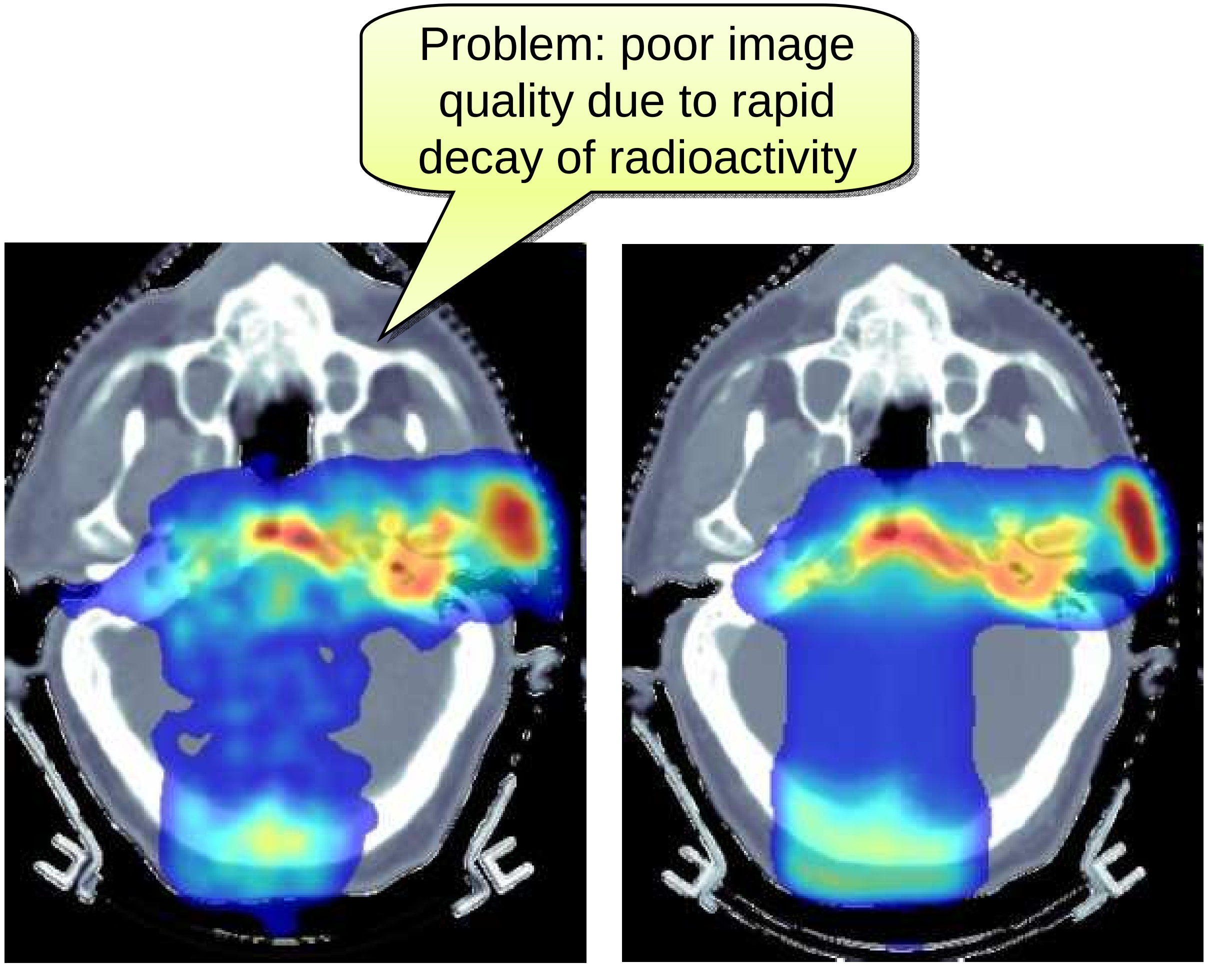
Prompt gamma rays:



Positron Emission Tomography



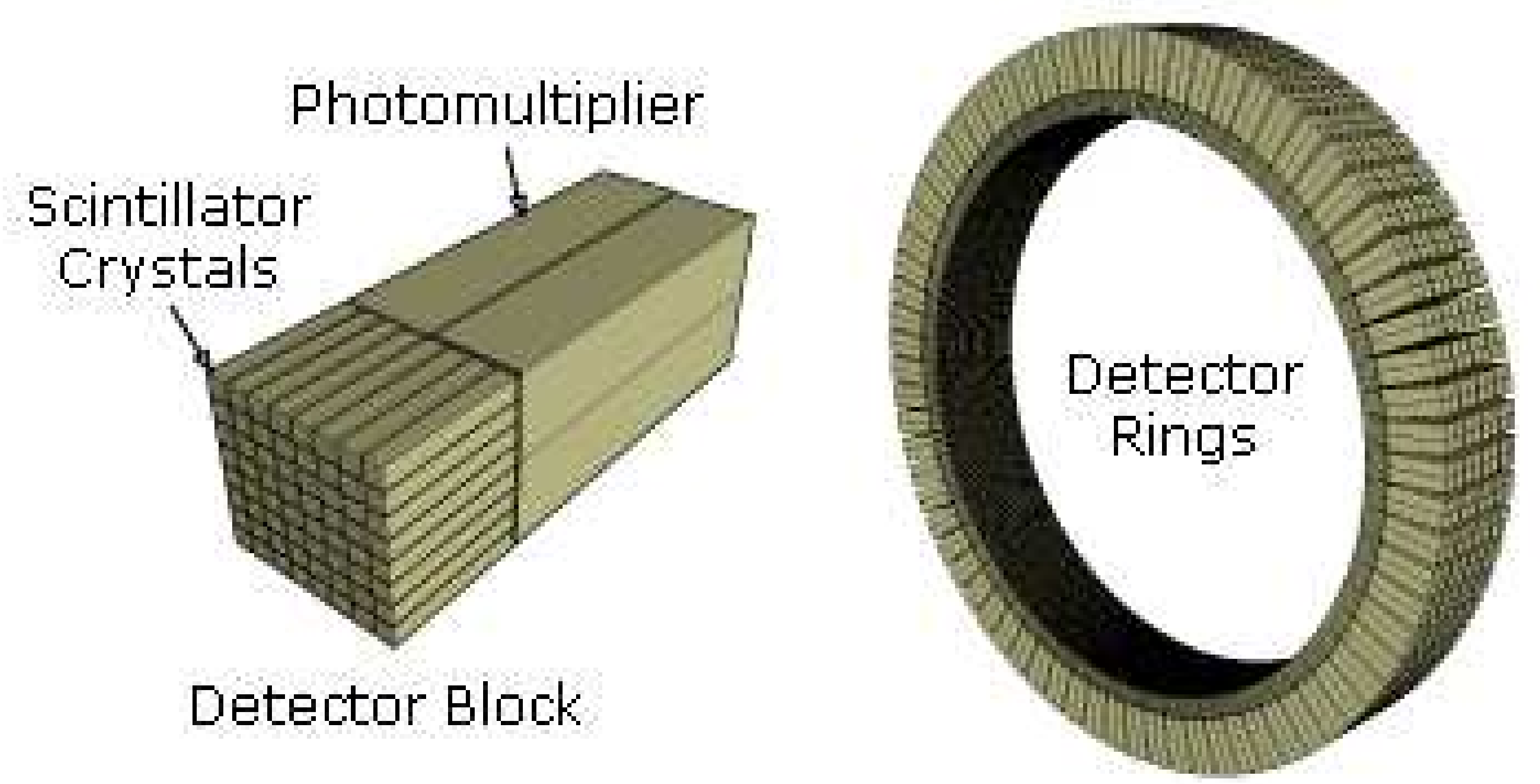
Offline PET



Commercial PET scanner

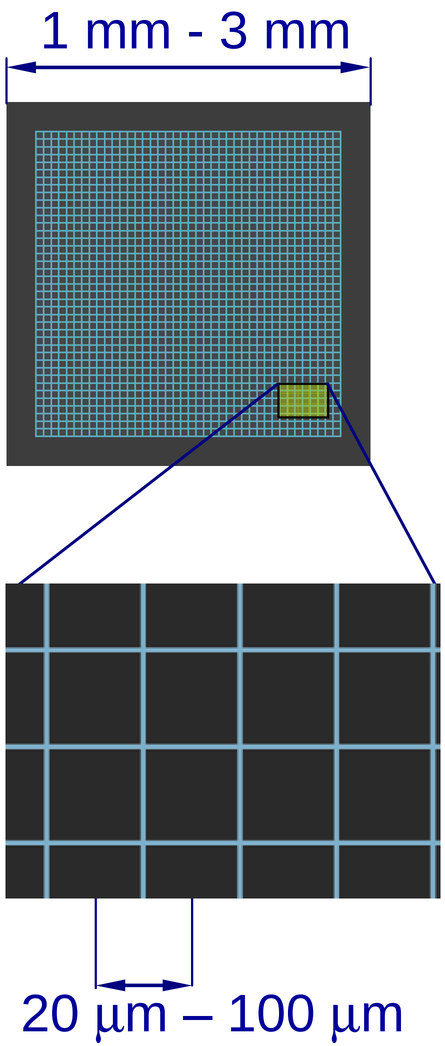
Activity distribution after 2-beam proton irradiation at Massachusetts General Hospital.
Right: predicted. Left: measured with PET scanner outside treatment room.

PET detectors: classic "block" detector



- Several block detectors are assembled into a ring
- A scanner may consist of several detector rings

Silicon Photomultiplier (SiPM)

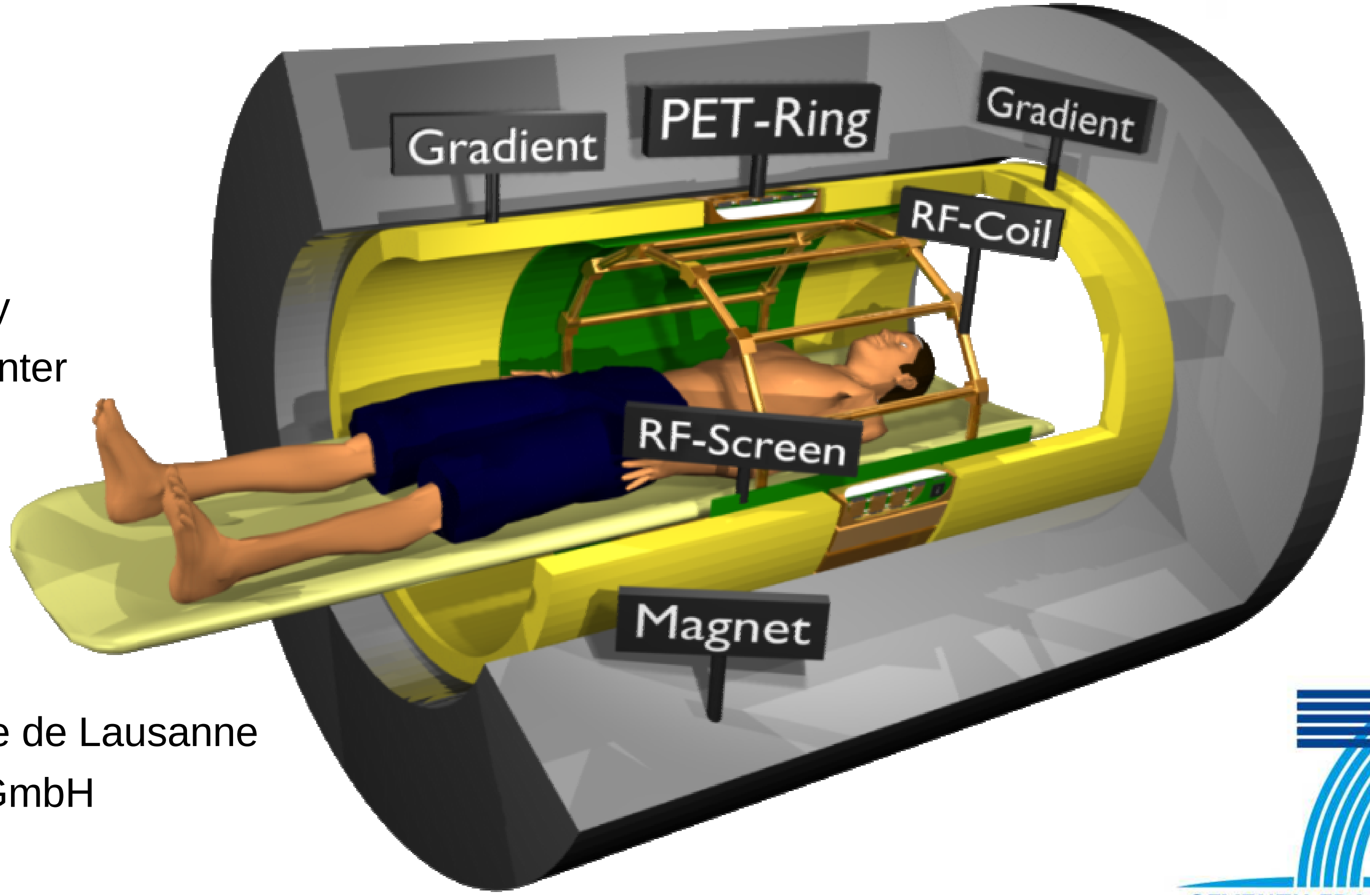


- Array of many self-quenched Geiger-mode APDs (microcells) connected in parallel
- Increasingly interesting as replacement for PMTs:
  - high gain (~10^6)
  - high PDE
  - compact and rugged
  - transparent to γ-photons
  - fast response (ns)
  - insensitive to magnetic fields

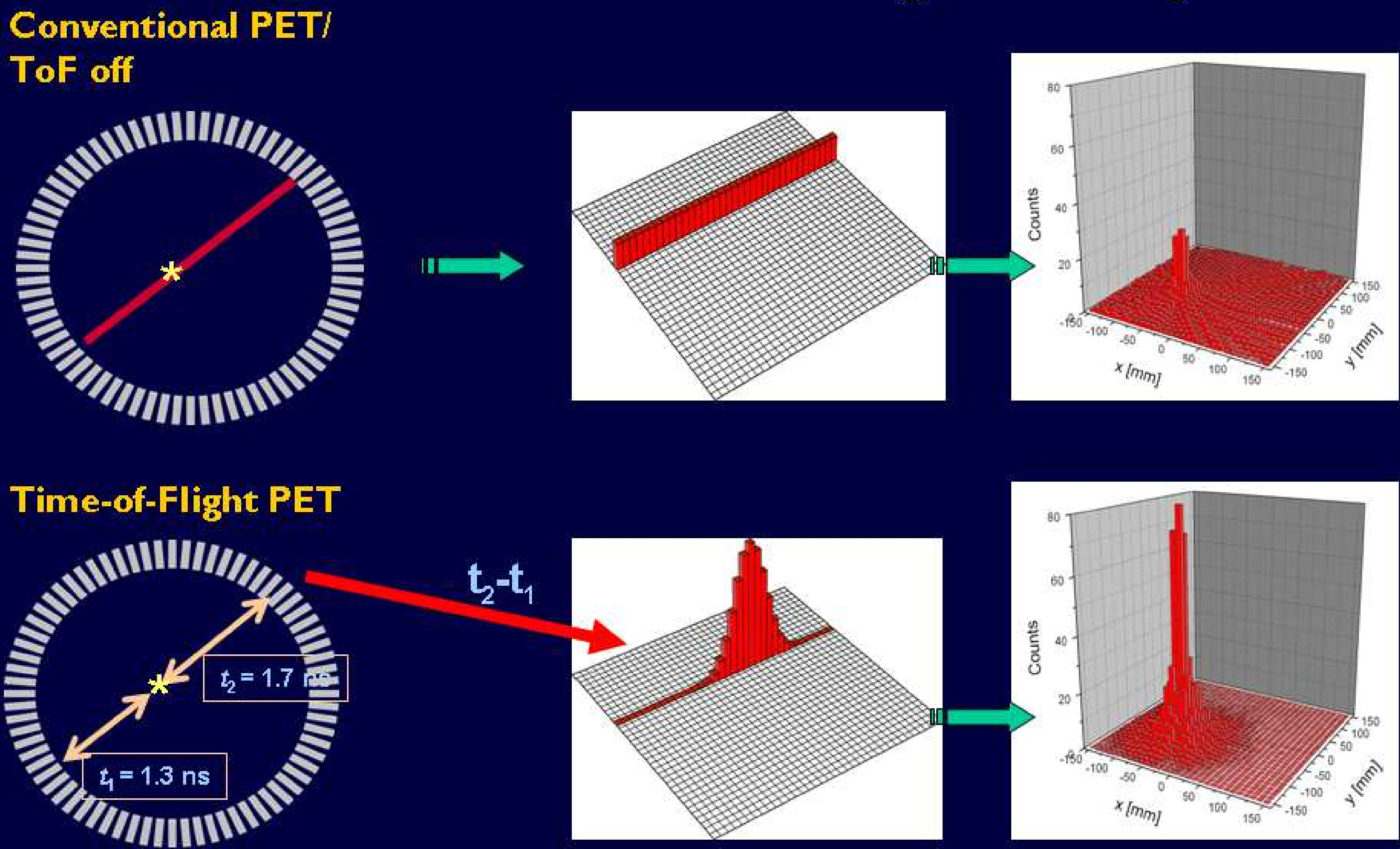
SUBLIMA Project

Whole-body TOF-PET / MRI

Philips Research
Delft University of Technology
Leiden University Medical Center
University of Heidelberg
University of Ghent
King's College London
Fondazione Bruno Kessler
University of Pennsylvania
Ecole Polytechnique Fédérale de Lausanne
Micro Systems Engineering GmbH
Technolution BV



Time of Flight PET Systems

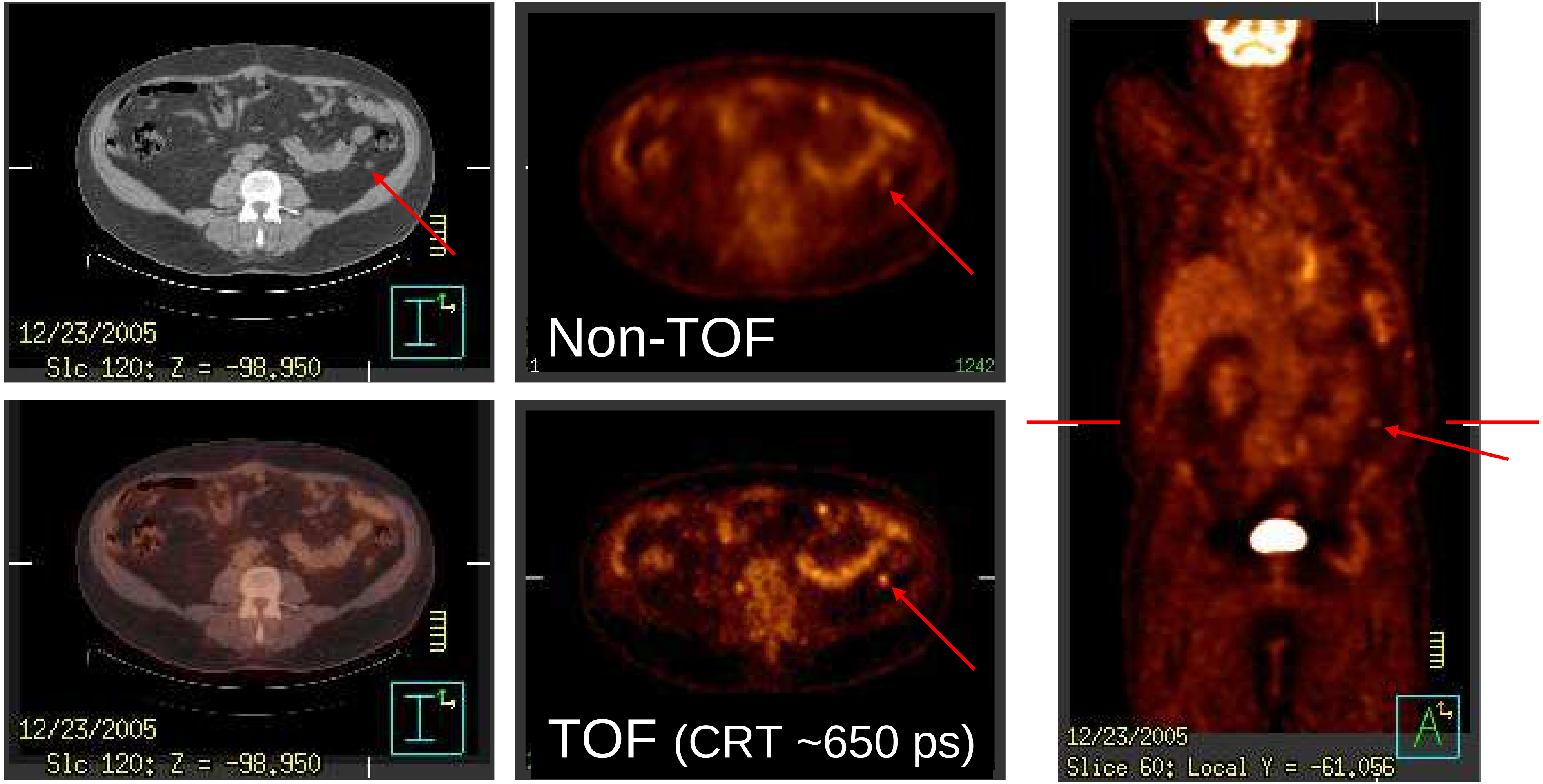


→ ToF: more signal, less noise

Time-of-flight (TOF) PET

114 kg; BMI = 32.2
13.4 mCi; 2 hr post-inj

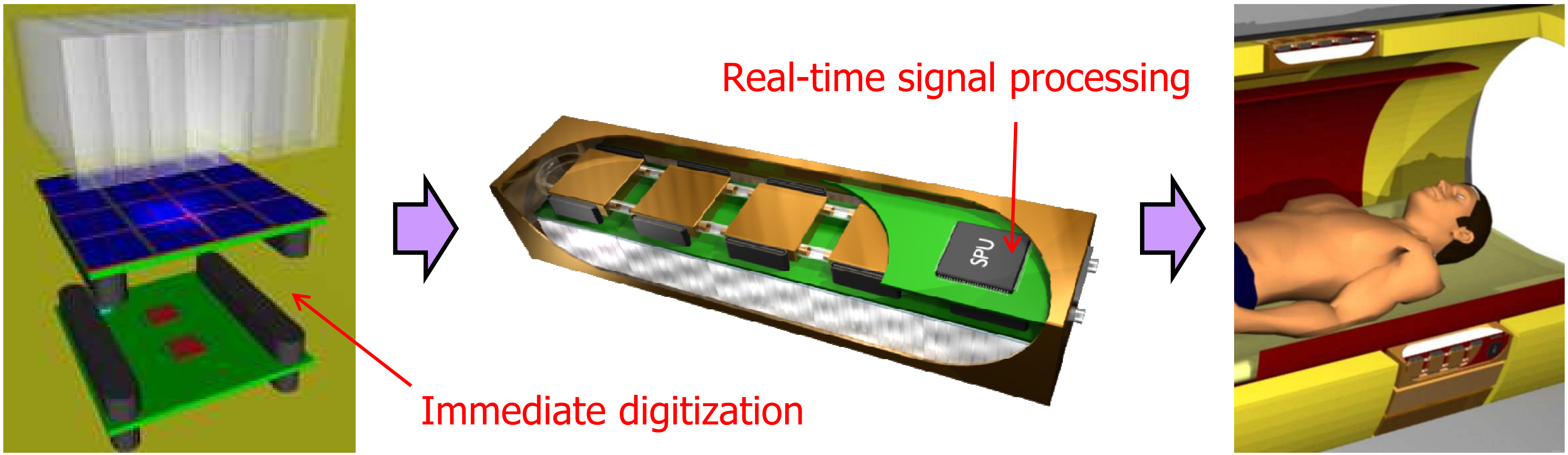
Colon cancer, left upper quadrant peritoneal node



State-of-the-art clinical PET: coincidence resolving time (CRT) ≈ 500 ps

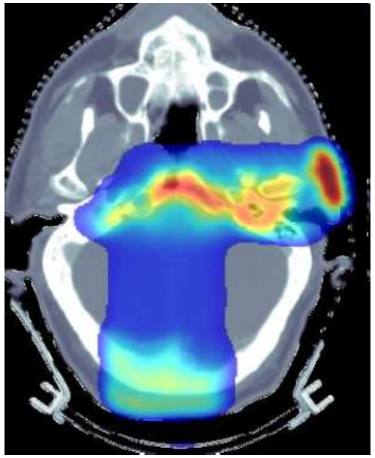
SiPM-based TOF-PET system

- Excellent time-of-flight performance
- MRI-compatible (up to 7 Tesla)
- High spatial resolution



Detector Module System

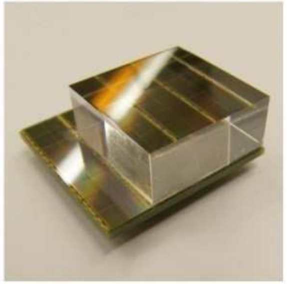
In-situ TOF-PET

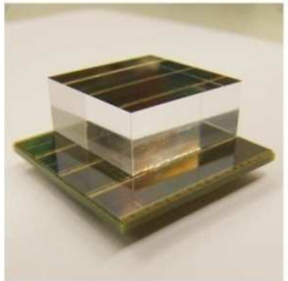


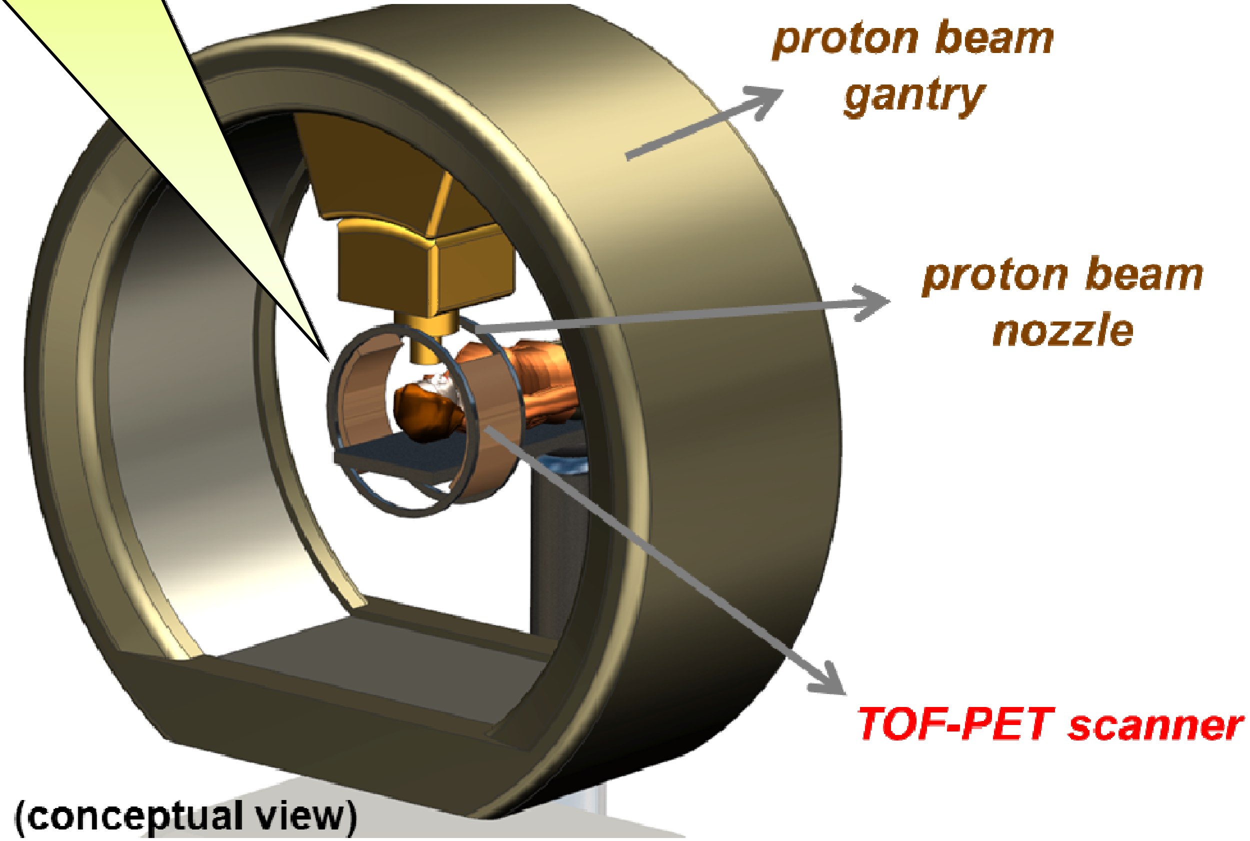
- **Avoid decay losses by in-situ imaging**
- **Use TOF for optimum image quality**

Incentive

Use revolutionary detection technology, under development for PET-MRI by TU Delft and Philips, to realize clinically useful in-situ dose imaging device

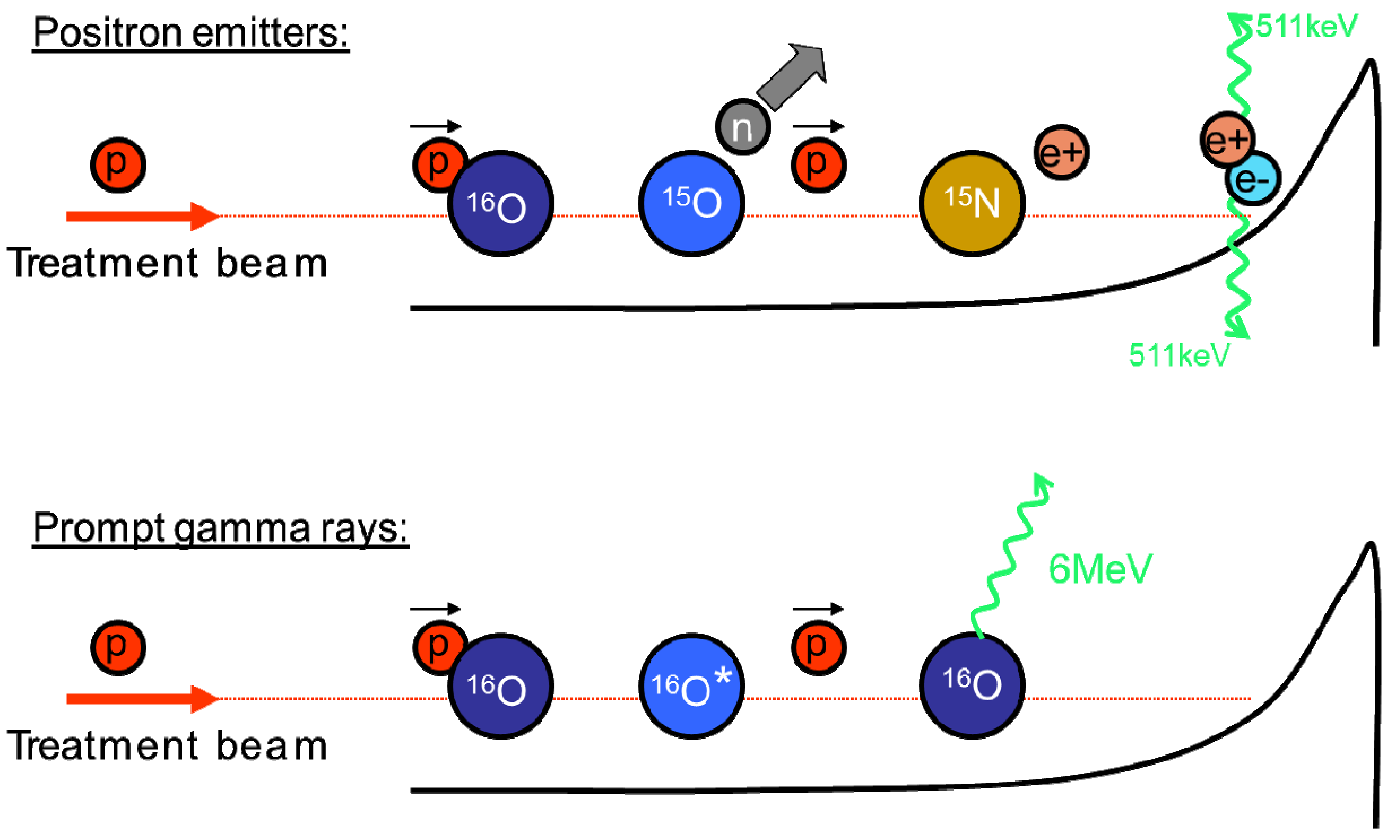




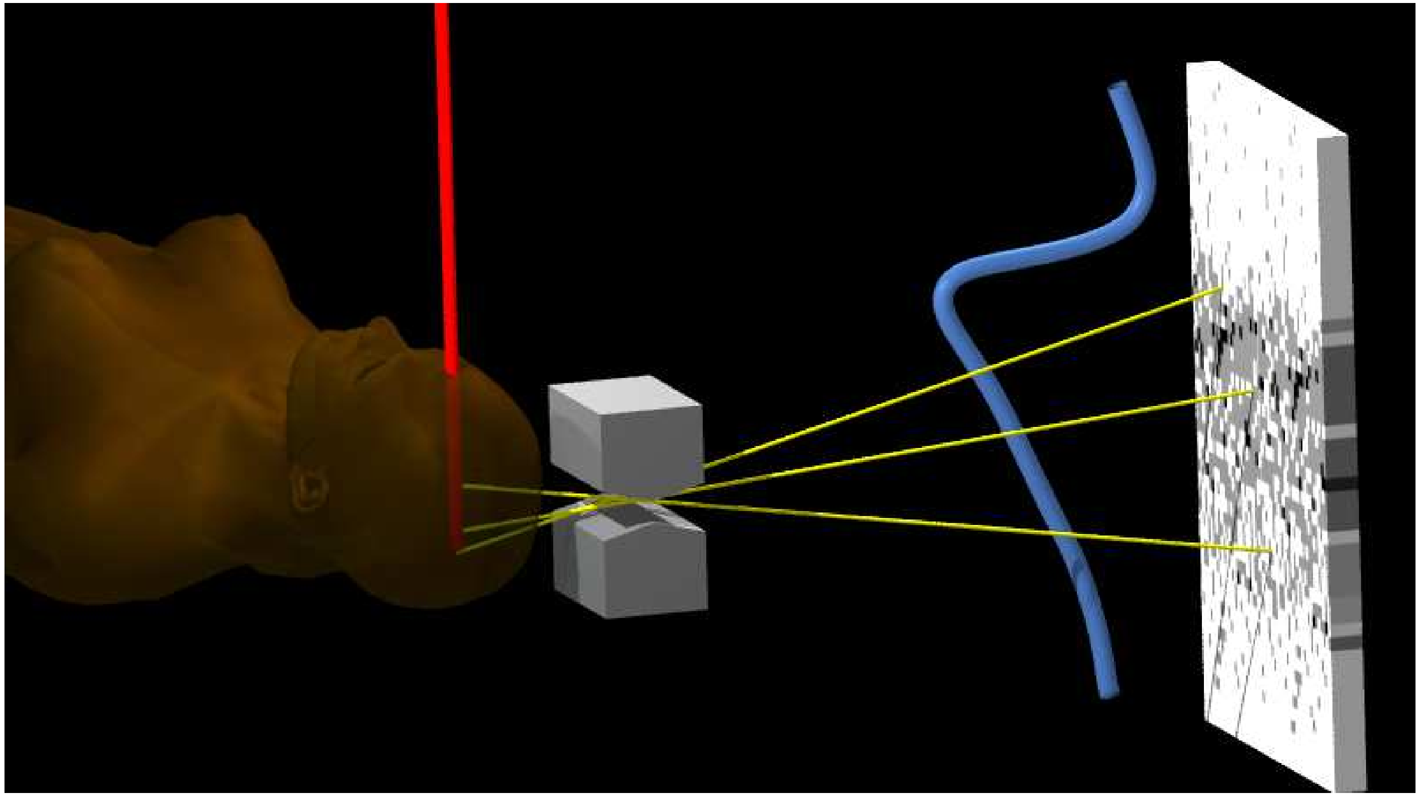


In-situ dose imaging

Human body > 90% Oxygen, Carbon, Hydrogen and Nitrogen

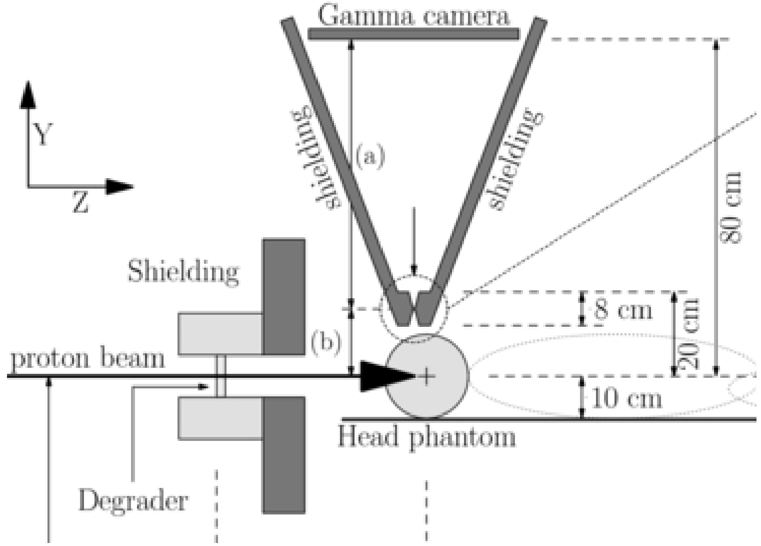


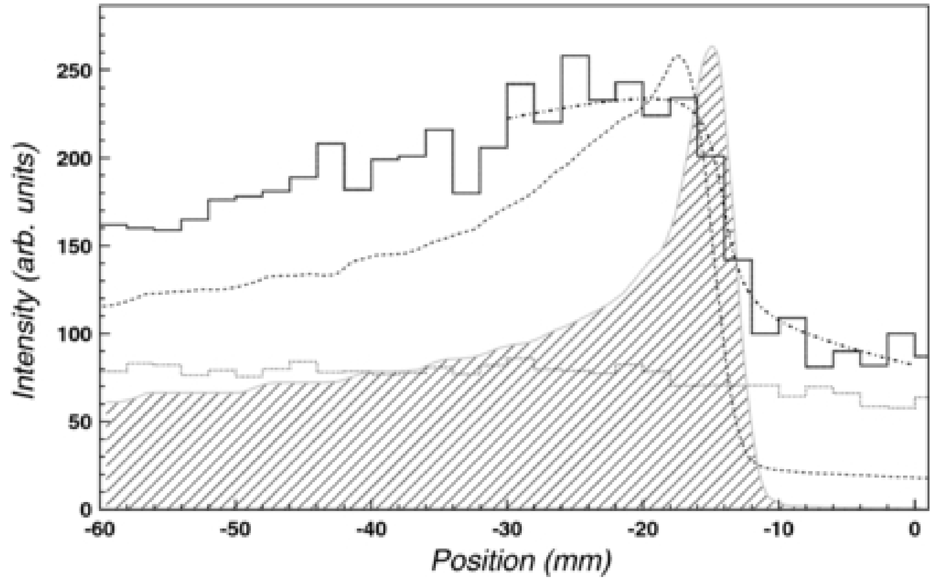
Prompt-gamma imaging (slit)



Slit camera for real time prompt gamma imaging for high precision proton therapy

Prompt-gamma imaging (slit)

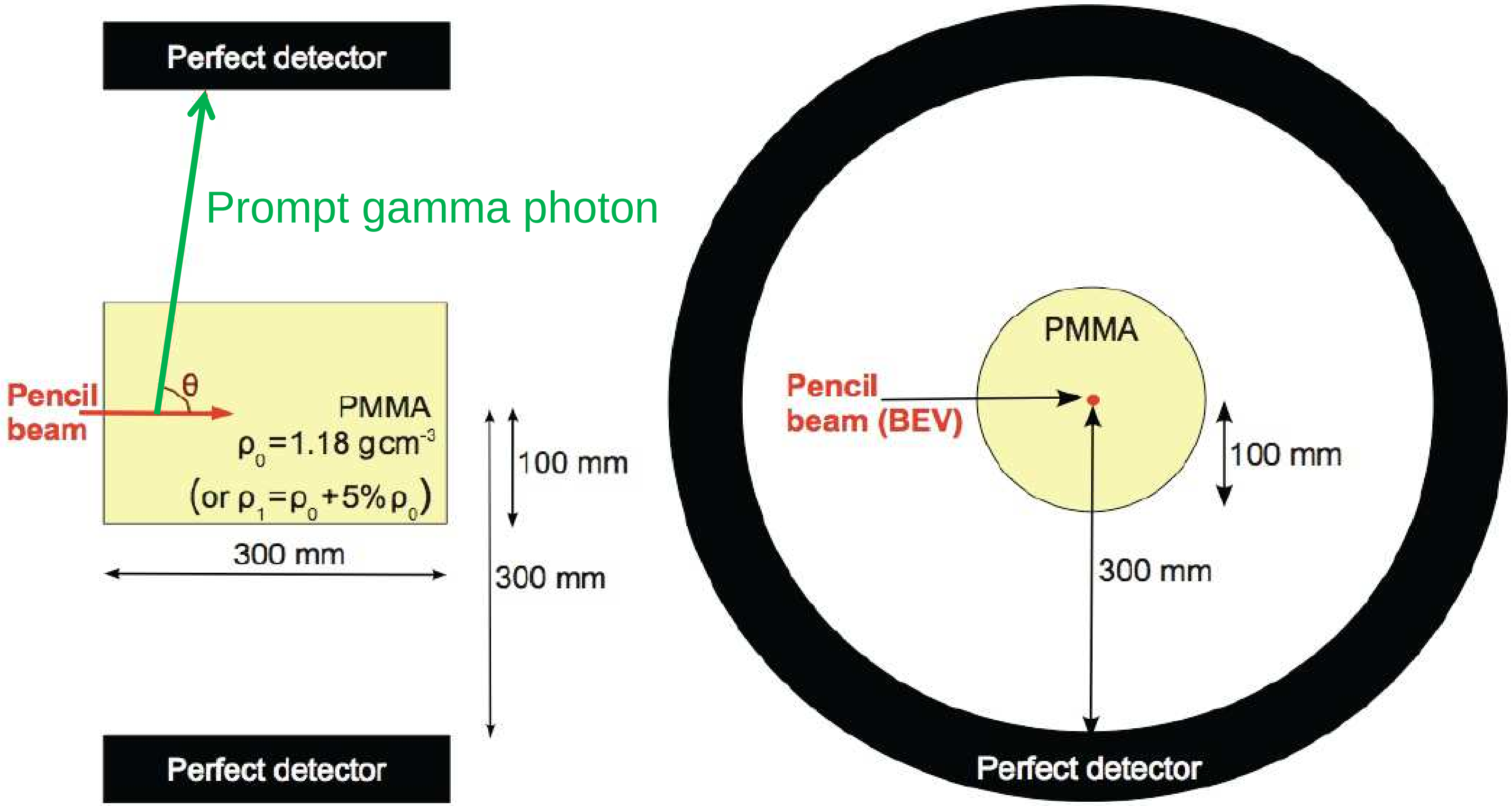




- Slit camera for real-time Bragg peak position verification in particle therapy
- Simulations indicate that under common therapy conditions enough data may be collected to accurately locate the distal dose edge during a spot-step
- This project aims to build the camera with optimal slit camera geometry and test it in particle therapy clinics

Prompt-gamma imaging (time-of-flight)

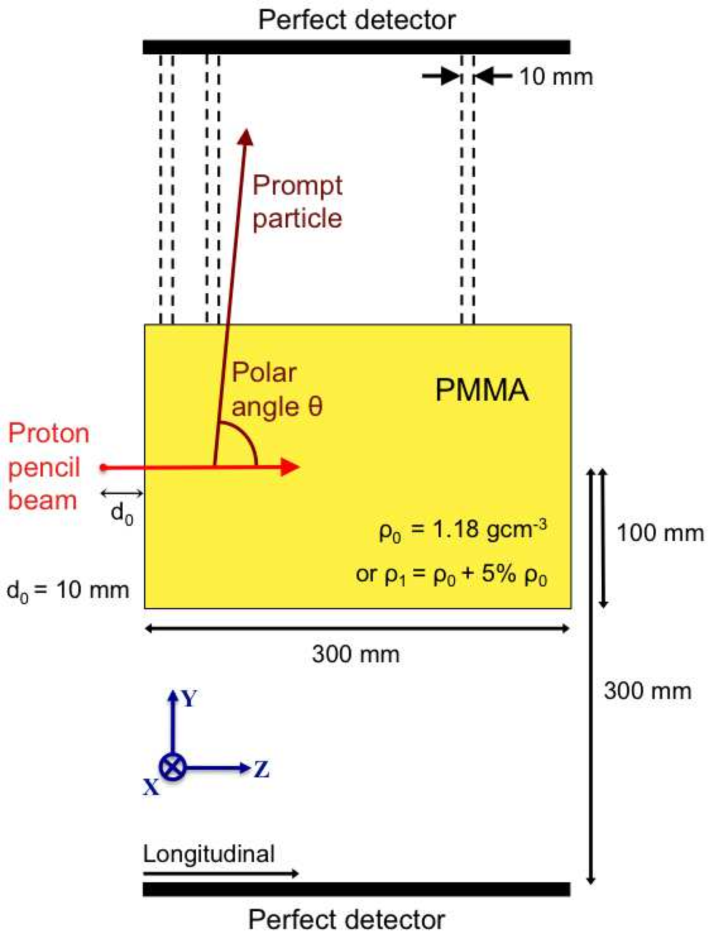
GEANT4 Monte Carlo simulation geometry

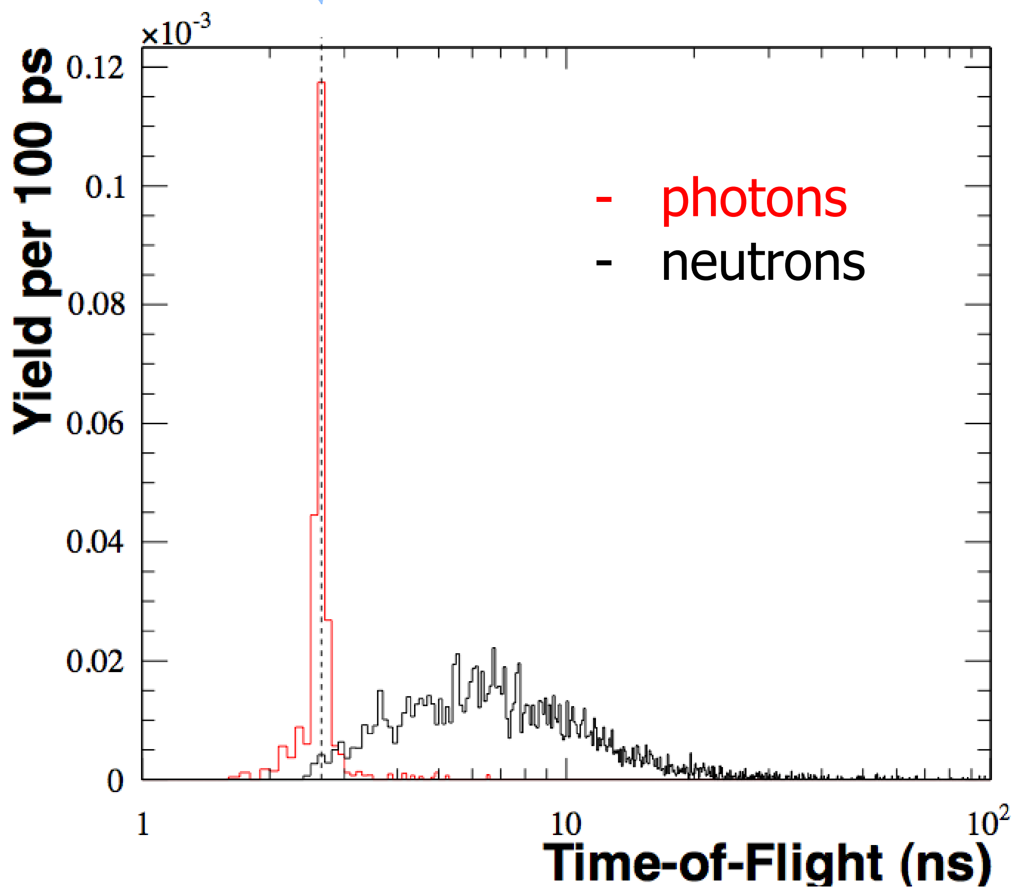


- Prompt-gamma's created by proton interactions escape from phantom
- Escaped gamma photons measured perpendicularly to the beam

Time-of-flight neutron rejection

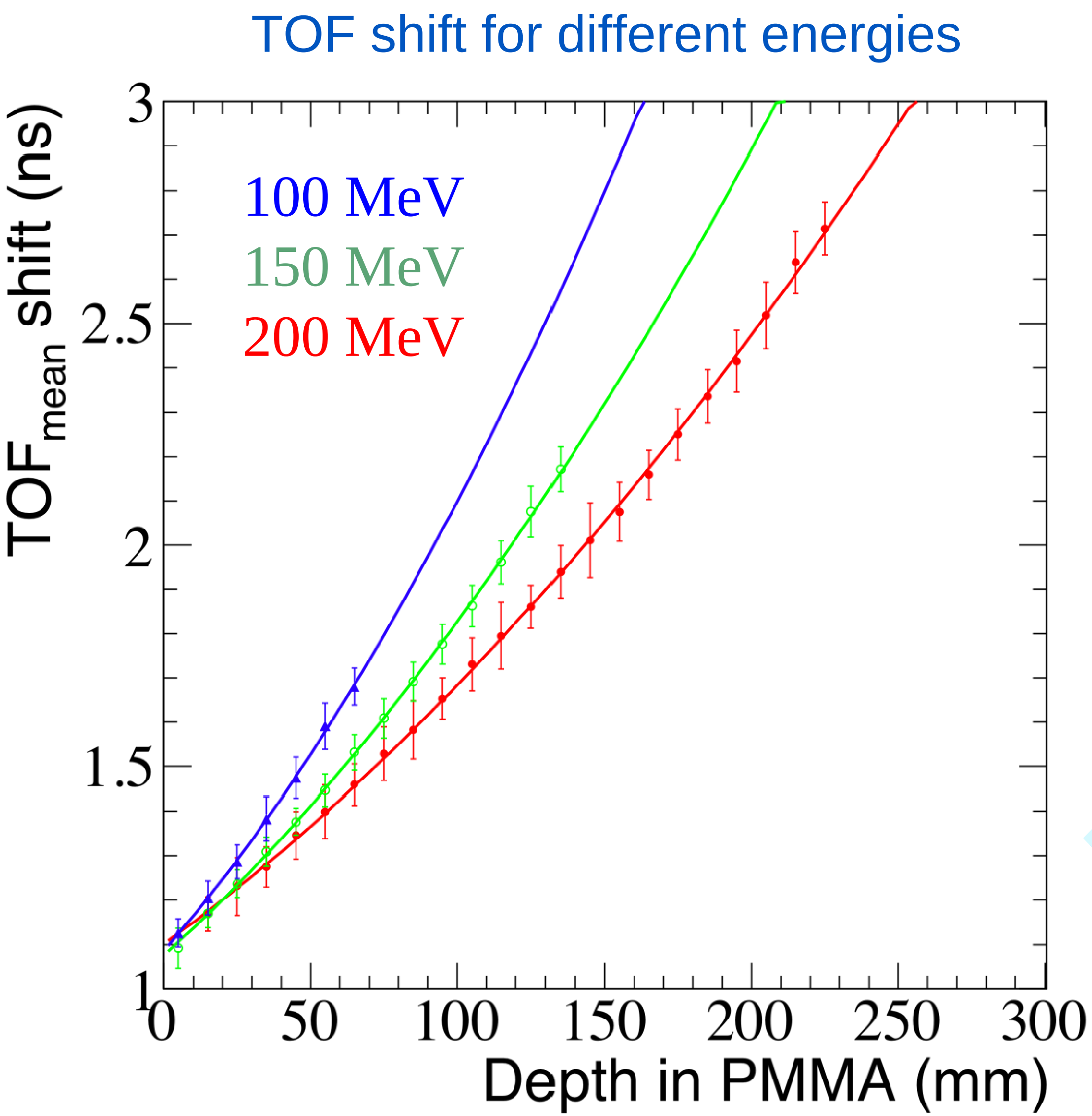
GEANT4 Monte Carlo simulation

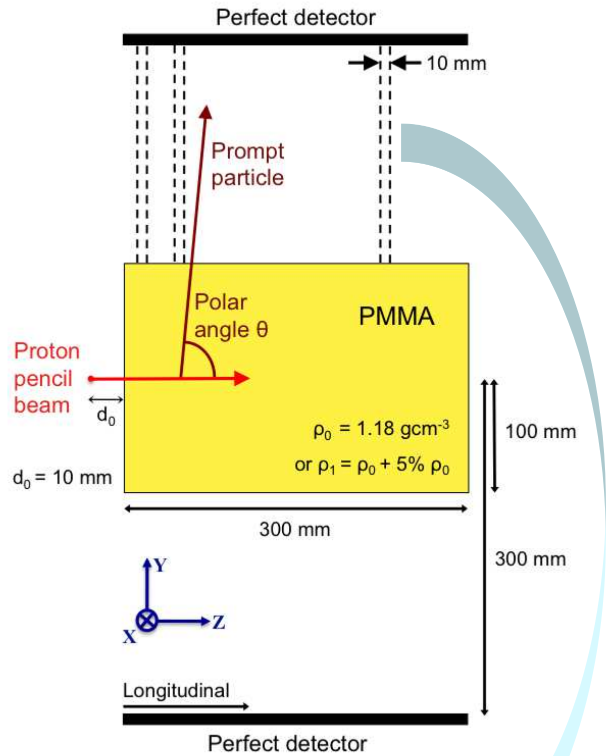


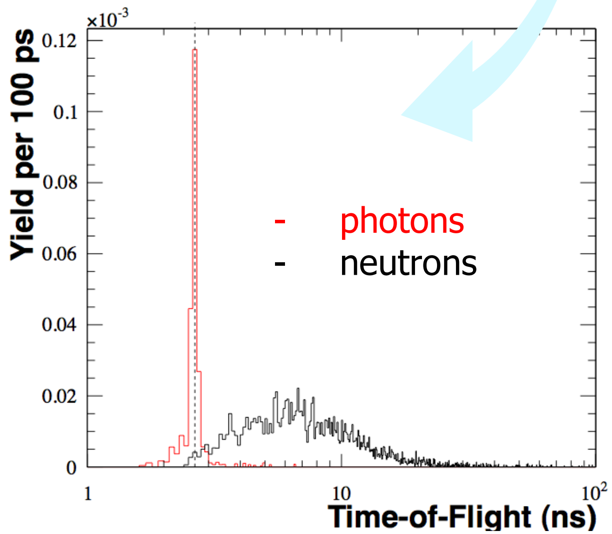


Biegun et al, Phys. Med. Biol. 57, 6429–6444 (2012)

TOF neutron rejection



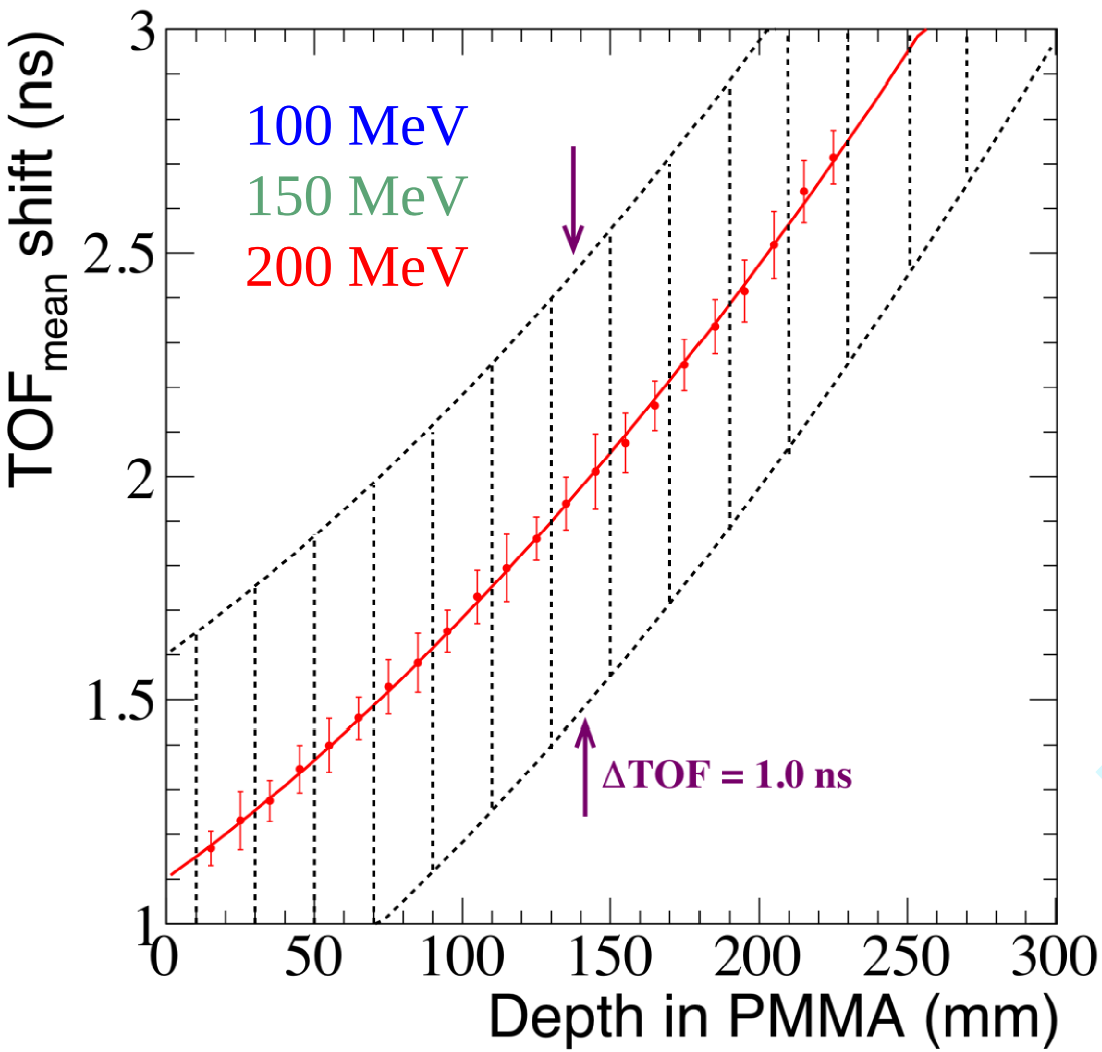


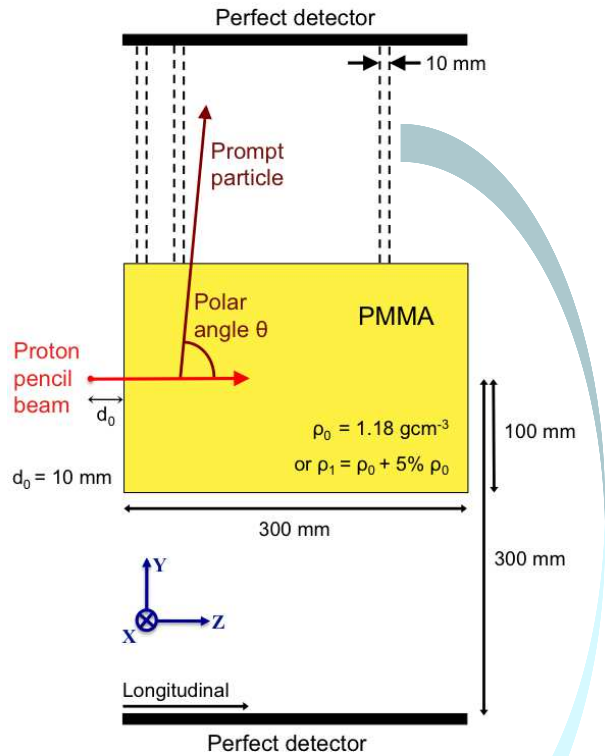


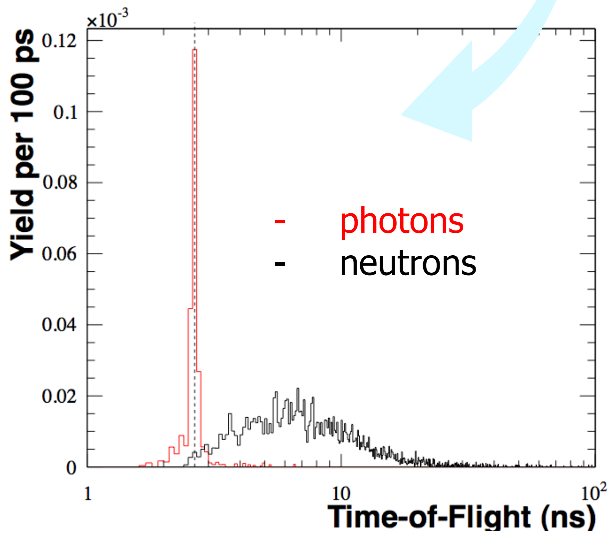
Biegun et al, Phys. Med. Biol. 57, 6429–6444 (2012)

TOF neutron rejection

TOF shift for different energies

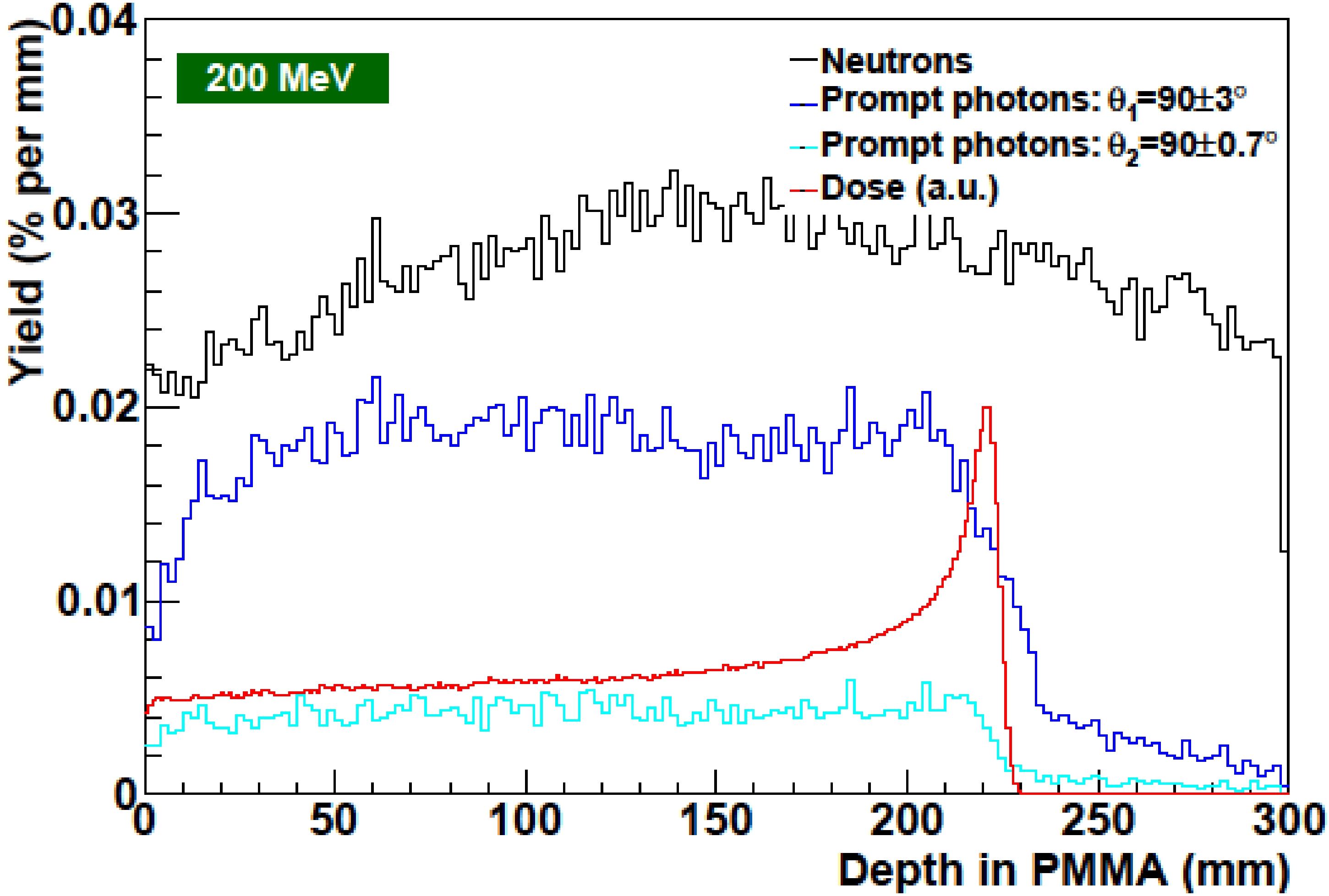






TOF neutron rejection

GEANT4 Monte Carlo simulation

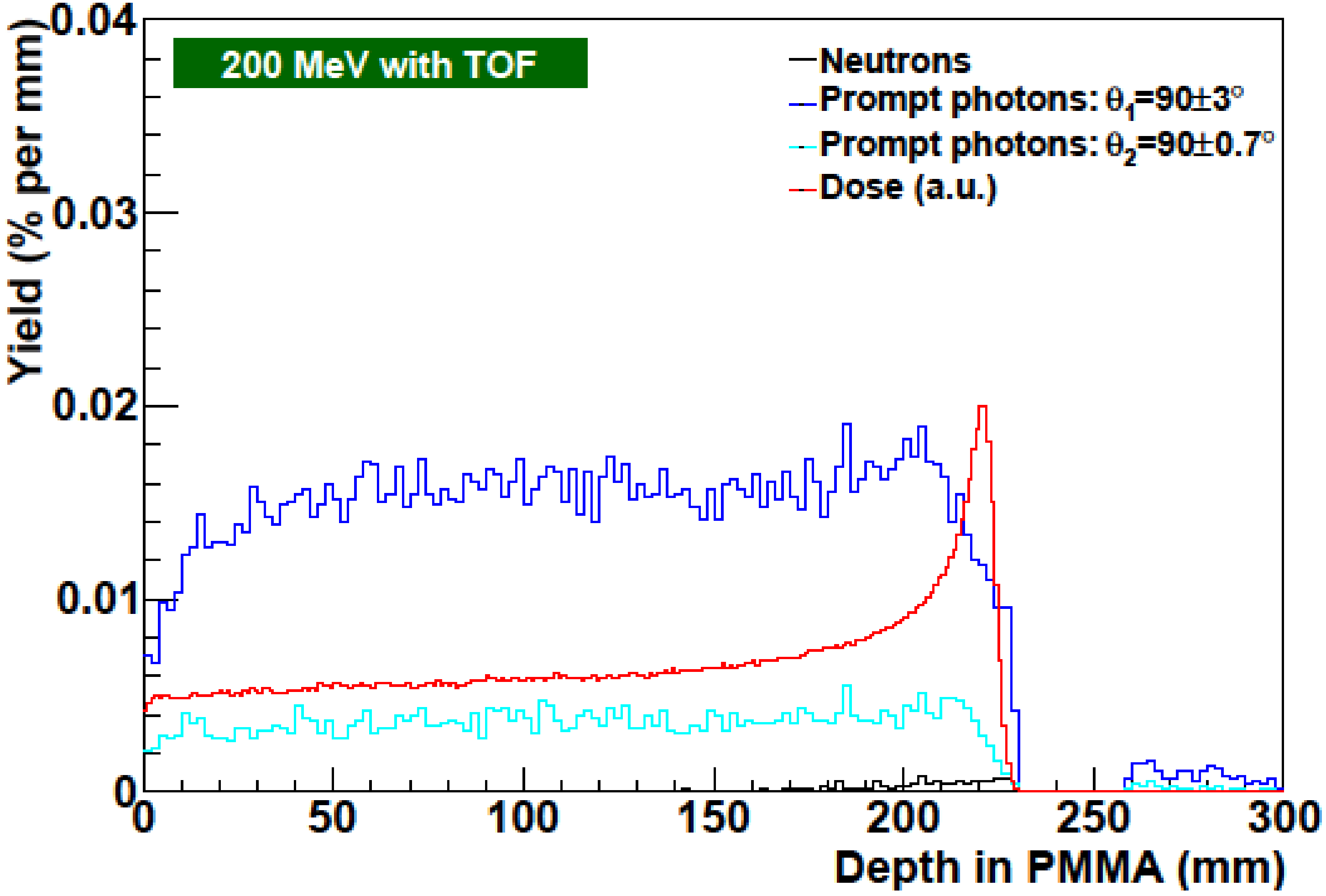


BEFORE

Biegun et al, Phys. Med. Biol. 57, 6429–6444 (2012)

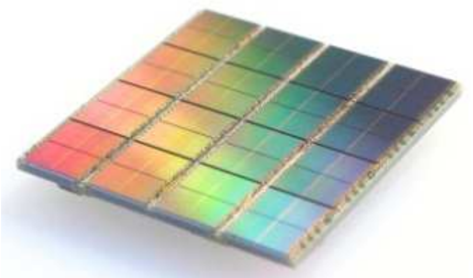
TOF neutron rejection

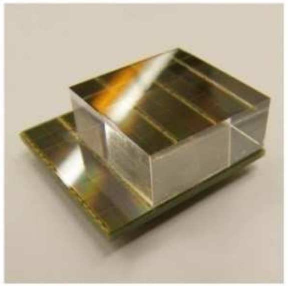
GEANT4 Monte Carlo simulation

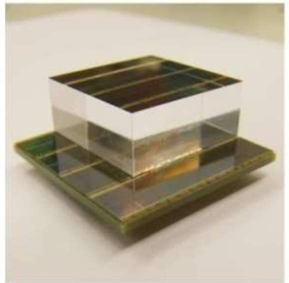


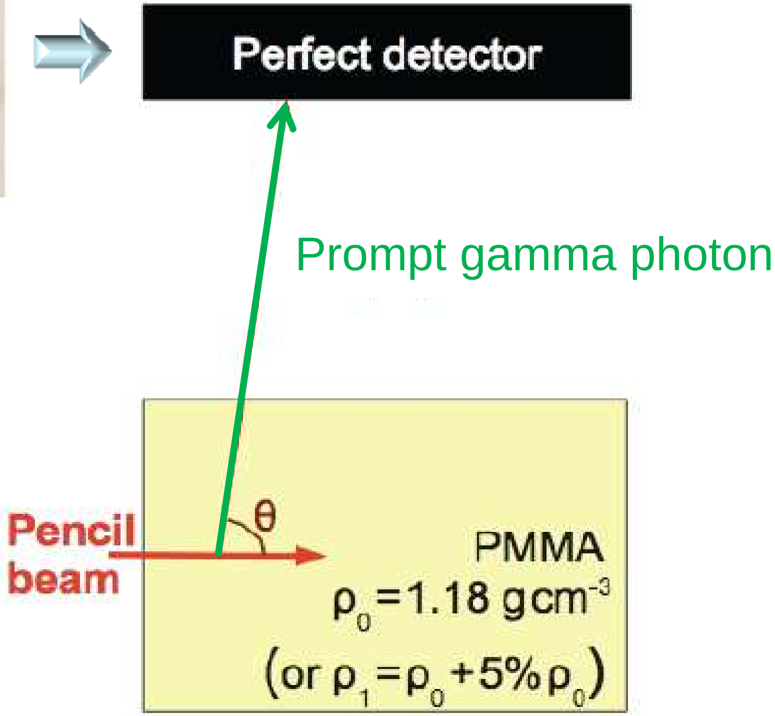
AFTER

TOF-PET detectors for PG imaging







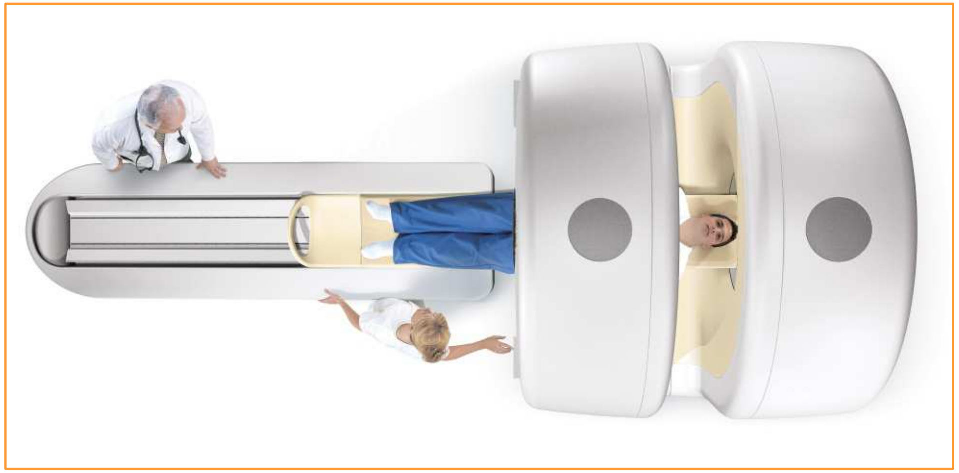


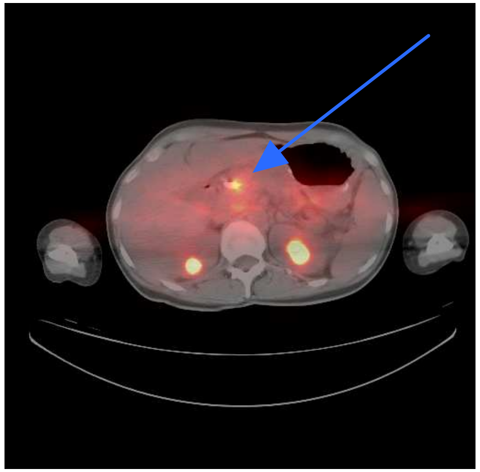
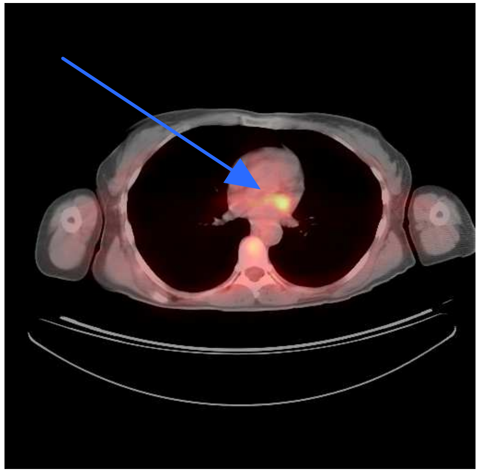
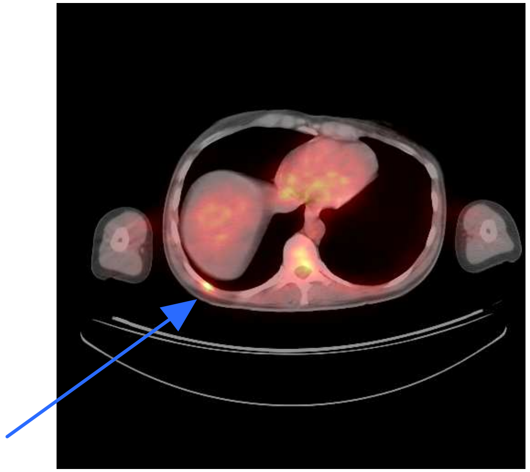
- High spatial resolution
- High count rate capability
- Time-of-flight ready
- Magnetic-field compatible
- Compact
- Scalable

Thank You

Backup slides

Clinical use: PET/CT



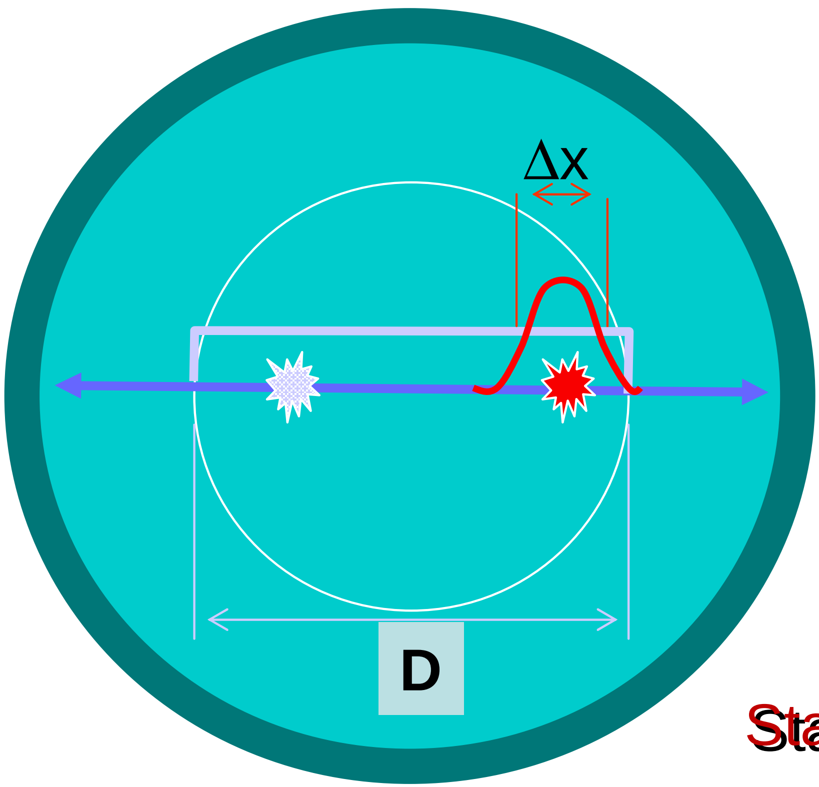




PET/CT (fused images): primary pancreatic cancer with suspicious chest wall and mediastinum lesions

Time-of-flight PET: concept of CRT

The accuracy of source position localization along line of response depends on the *coincidence resolving time (CRT)*

Δx = uncertainty in position along LOR = $c \cdot$ CRT/2, where c is the speed of light.



The TOF benefit is proportional to $\Delta x/D$, where D is the effective patient diameter.

=> The smaller the CRT, the better.

State-of-the-art: CRT $\approx$ 500 ps $\Rightarrow$ $\Delta x \approx$ 7.5 cm.

Dennis R. Schaart
Delft University of Technology

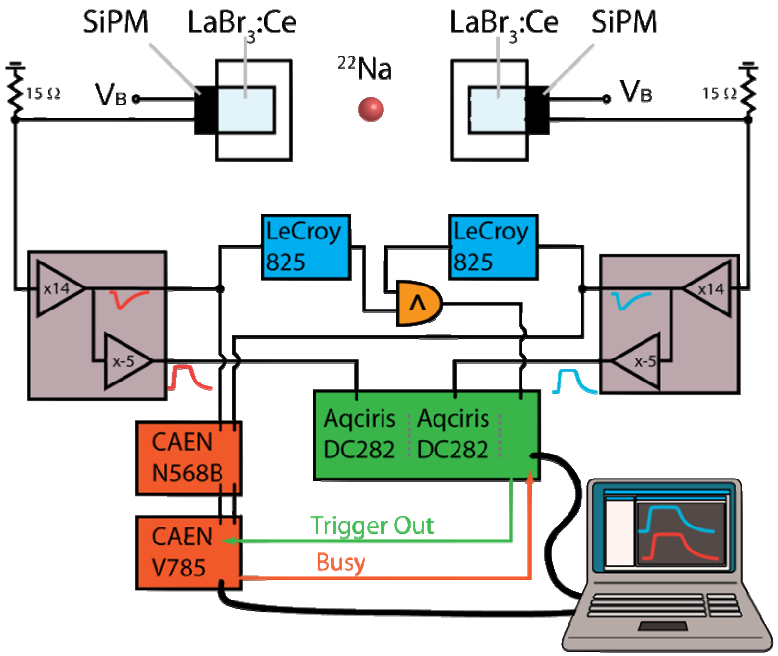
100 ps barrier broken using SiPMs

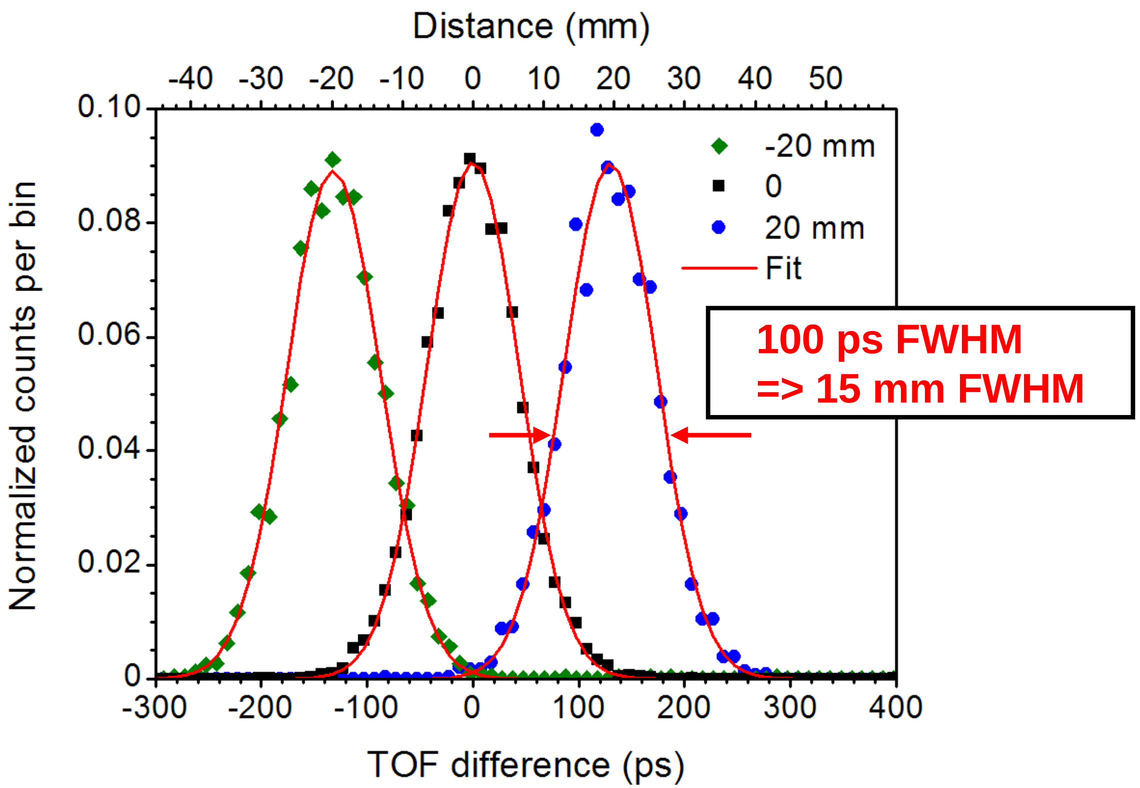
Made possible by the combination of:

•Small $LaBr_3$:Ce(5%) crystals (3 mm x 3 mm x 5 mm)

•Silicon Photomultipliers (Hamamatsu MPPC-S10362-33-050C)

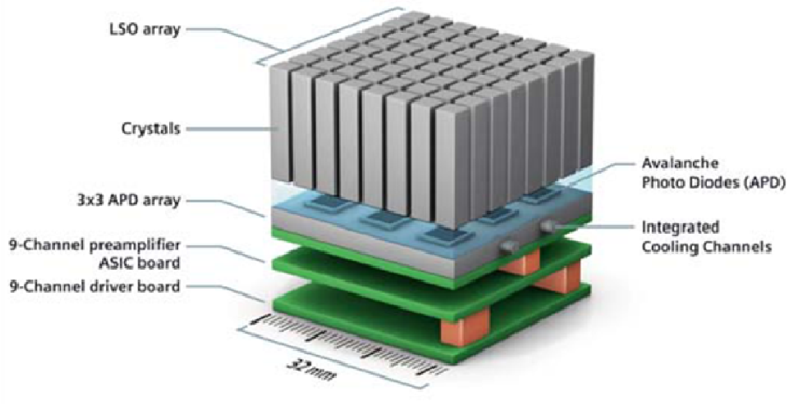
•Digital Signal Processing (DSP)





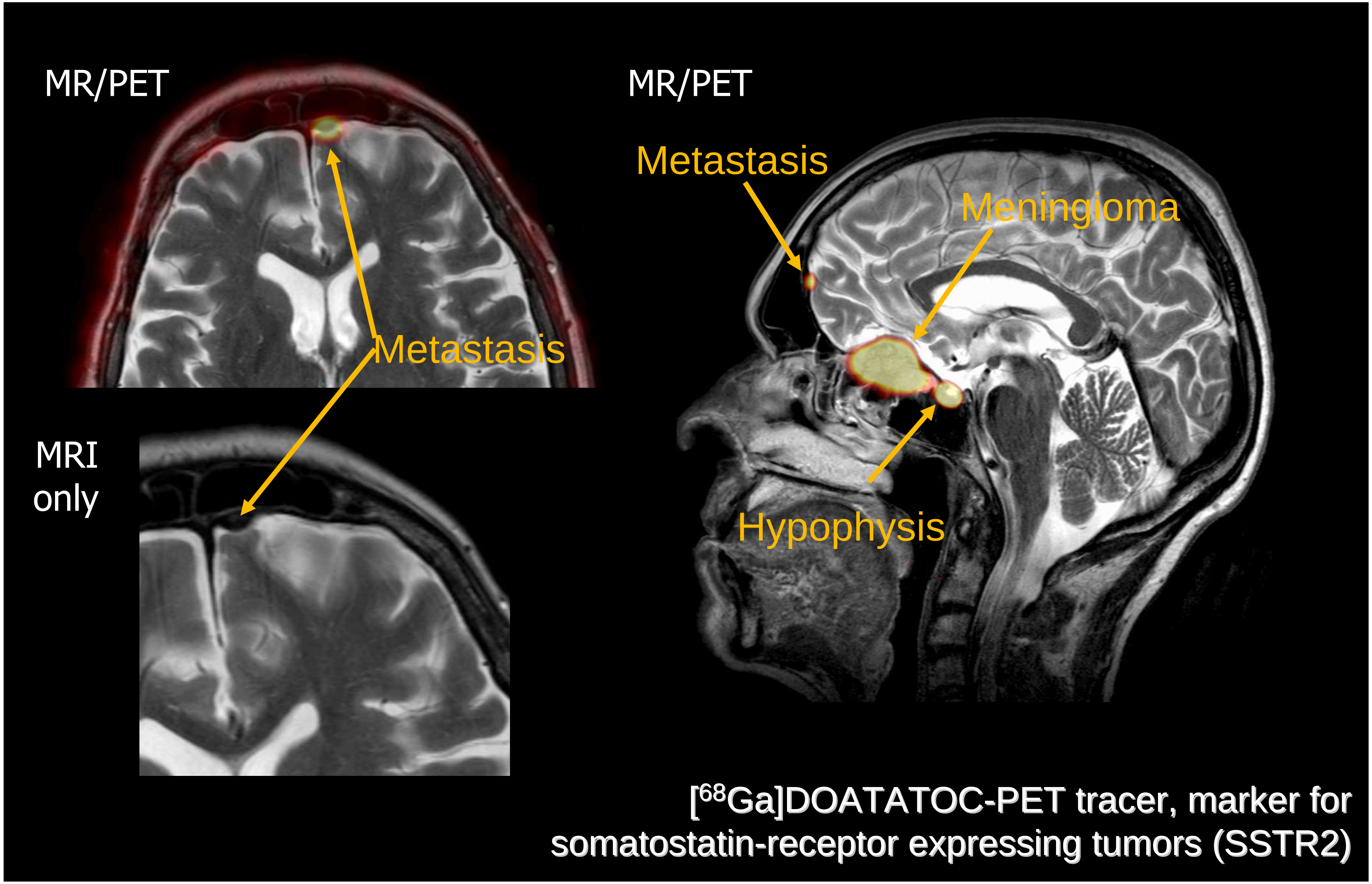
Multimodality: PET + MRI

Now: avalanche photodiodes (APDs)
Next generation systems: SiPMs

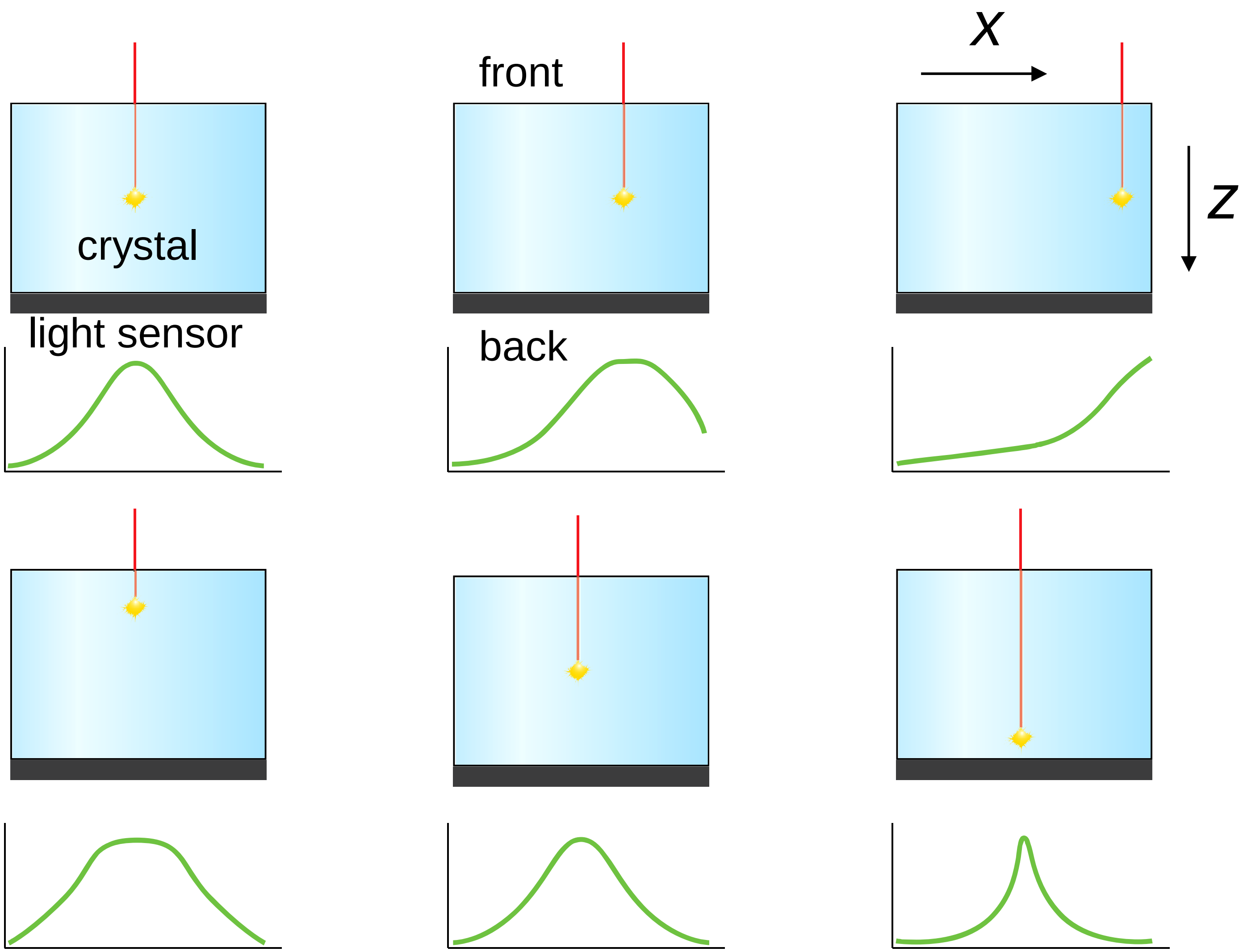


Images: Siemens

Multimodality: PET + MRI



Monolithic scintillator detectors

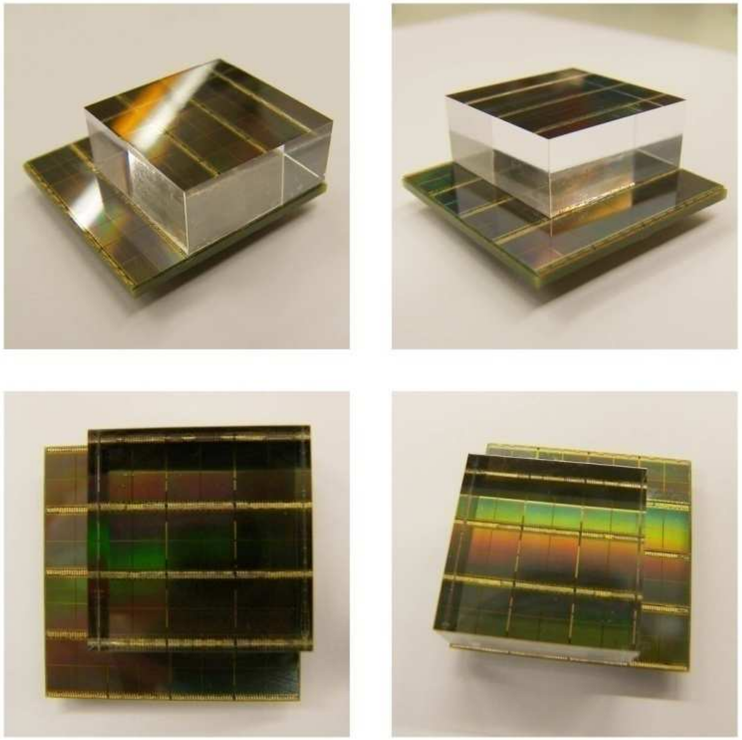


Light distribution depends on the entry point on the front surface…

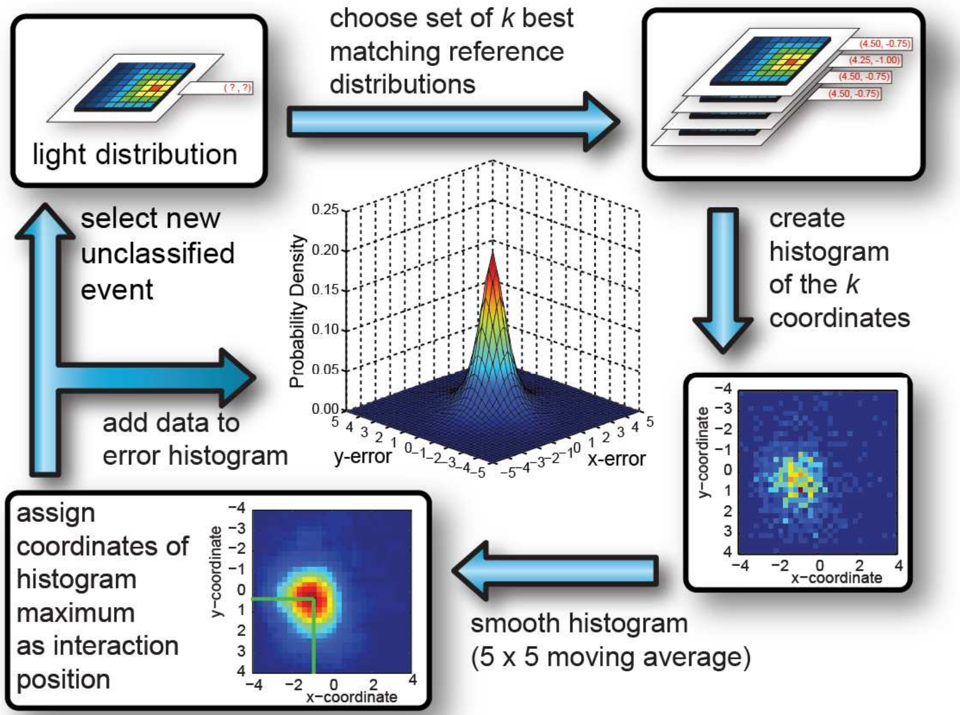
and on the depth of interaction (DOI).

dSiPM based monolithic scintillator

Monolithic TOF/DOI detector with improved performance due to Ca co-doped LSO scintillator, digital photon counting (dSiPM), and optimized readout algorithms

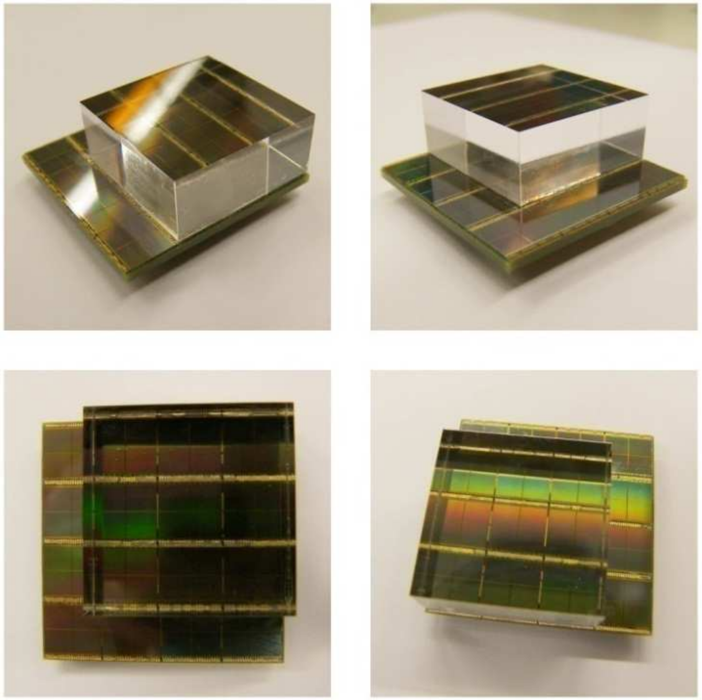


24 mm x 24 mm x 10 mm LSO:Ce,Ca scintillator on PDPC digital SiPM array



Faster & more accurate nearest-neighbour algorithm, H.T. van Dam et al, IEEE Trans Nucl Sci 58, 2139-2147, 2011

High-resolution, TOF & DOI !



Summary of first results with LSO:Ce,Ca monolithic scintillators on digital SiPM arrays:

- ~1 mm FWHM resolution (height = 10 mm)
- ~1.5 mm FWHM resolution (height = 20 mm)
- Coincidence resolving time ≤ 200 ps FWHM
- 11% - 12% FWHM energy resolution
- Intrinsic depth-of-interaction (DOI) information

$\Rightarrow$ A highly promising detector concept for clinical PET/CT and PET/MRI