
CERN

Principle of PET and some examples of PET developed within the Crystal Clear Collaboration

Dr. Marco Pizzichemi
CERN

Outline

- Introduction
- The PET system and its challenges
- PET in the Crystal Clear Collaboration
  - ClearPET
  - ClearPEM-Sonic
  - EndoTOFPET-US

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Positron Emission Tomography

- Definition:
  - In vivo imaging technique to quantitatively measure the 3D distribution of radiolabeled biomolecules

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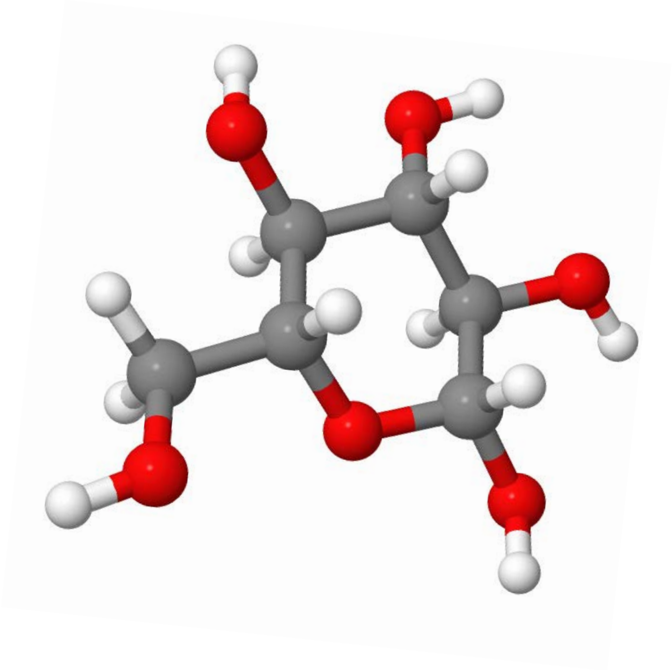
- Definition:
  - In vivo imaging technique to quantitatively measure the 3D distribution of radiolabeled biomolecules
- Applications:
  - Oncology
  - Neurology
  - Cardiology
  - Drug development
  - More…

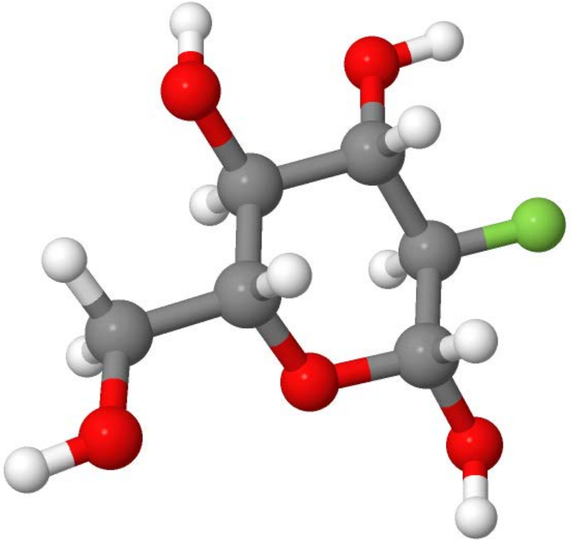
Positron Emission Tomography



Carbon
Oxygen
Hydrogen

Example of biomolecule: **glucose**

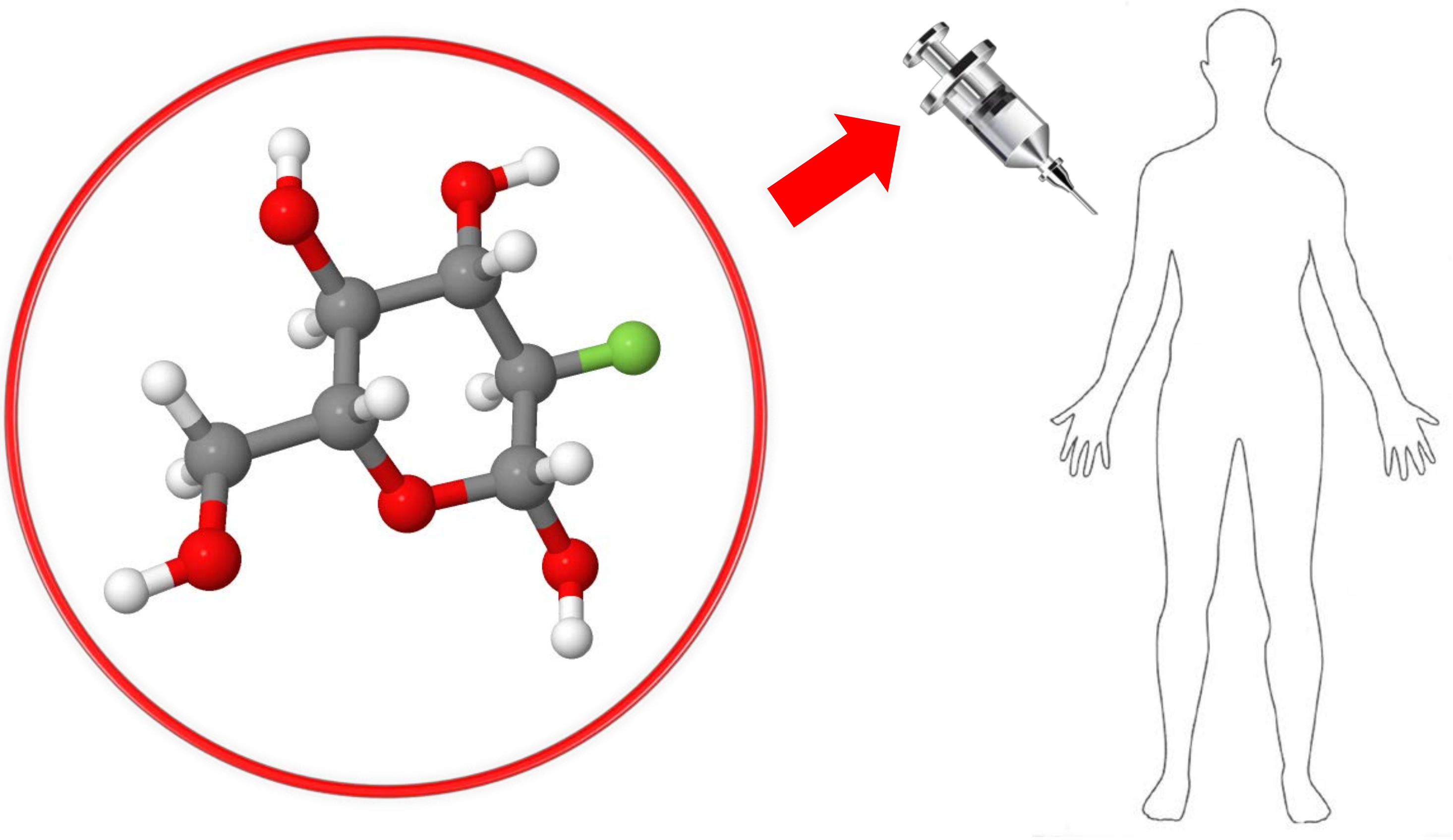
Positron Emission Tomography



Carbon
Oxygen
Hydrogen
Fluorine 18

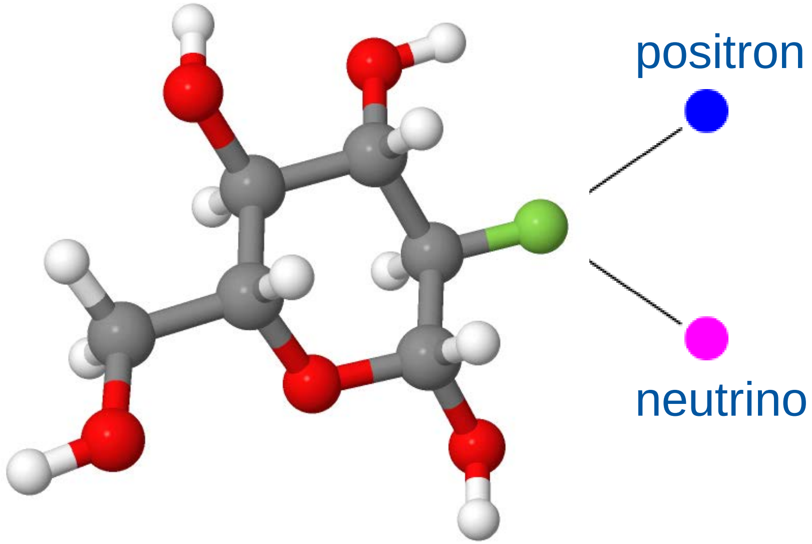
OH^{-} group replaced by ^{18}F: Fludeoxyglucose (**FDG**)

Positron Emission Tomography



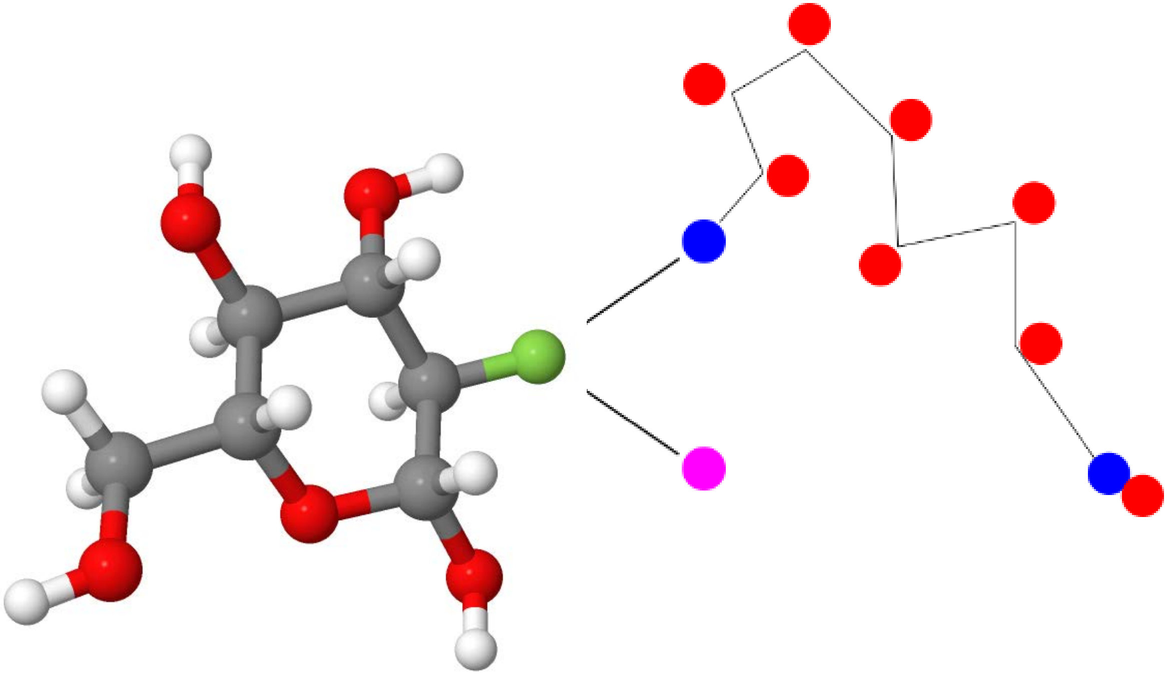
Injection in the patient

Positron Emission Tomography



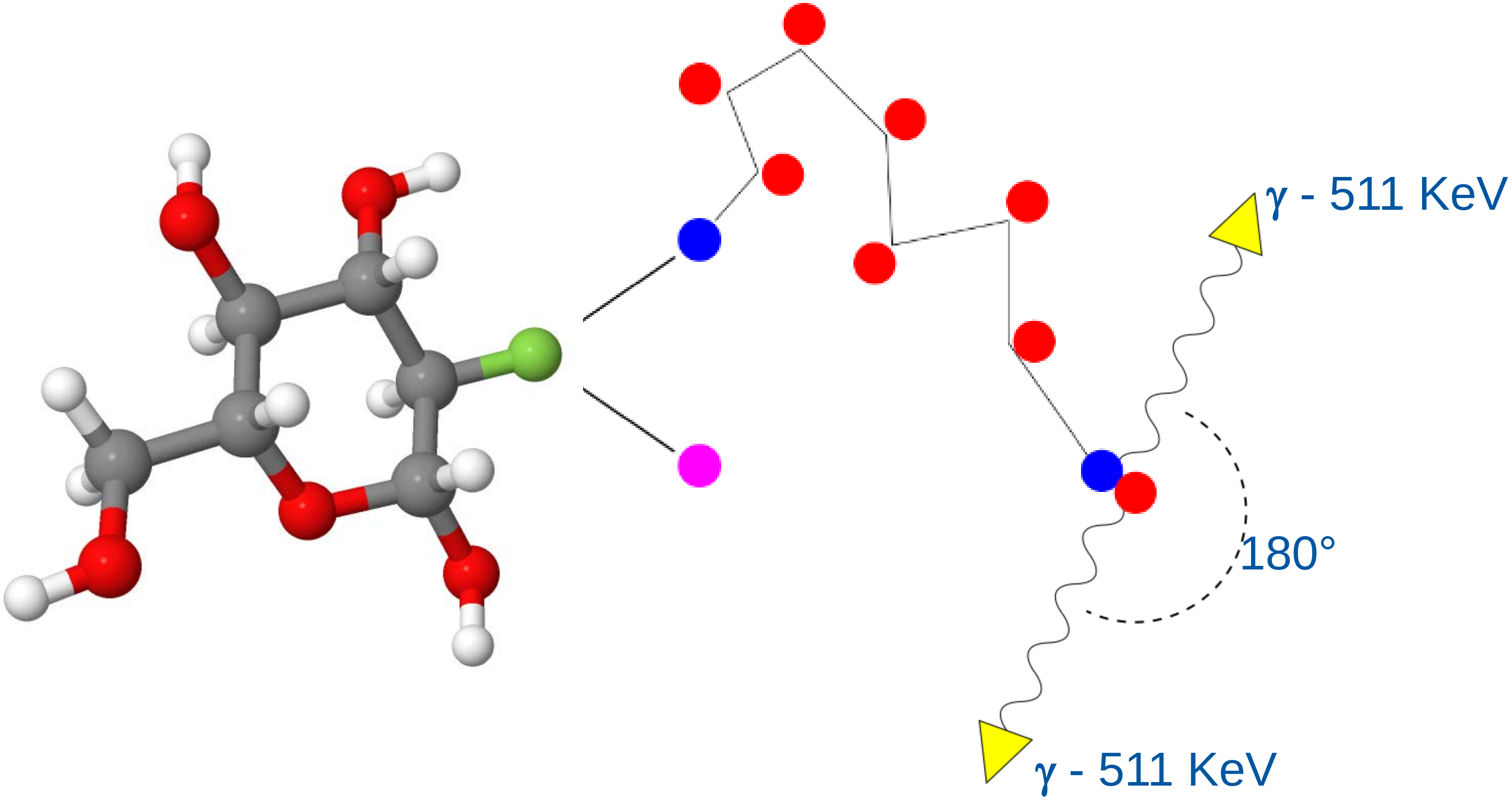
^{18}F decays emitting a **positron** and a **neutrino**

Positron Emission Tomography



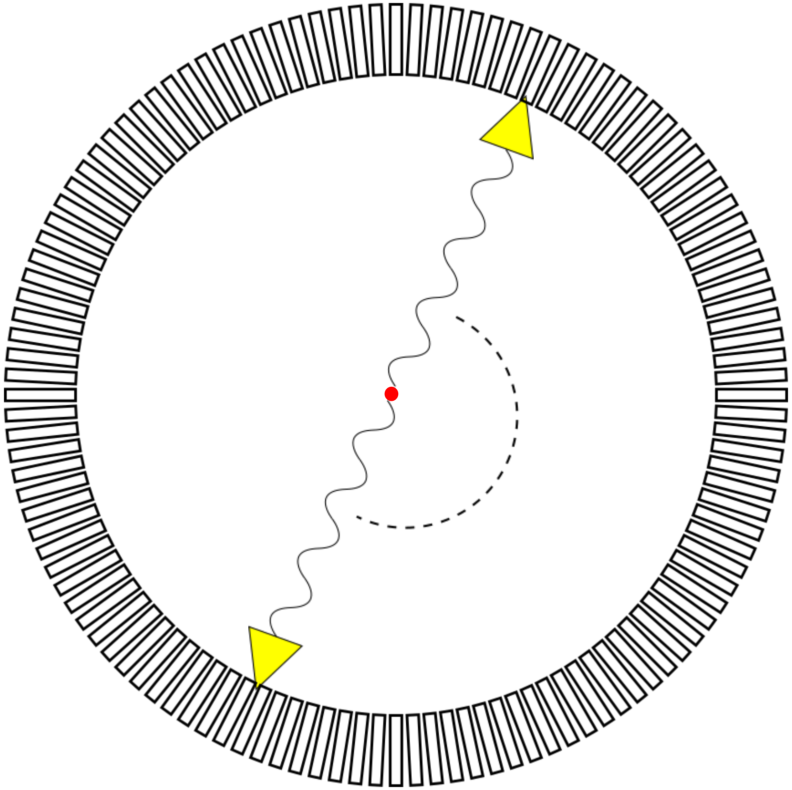
The positron travels in the tissue until it **annihilates** with an **electron**

Positron Emission Tomography



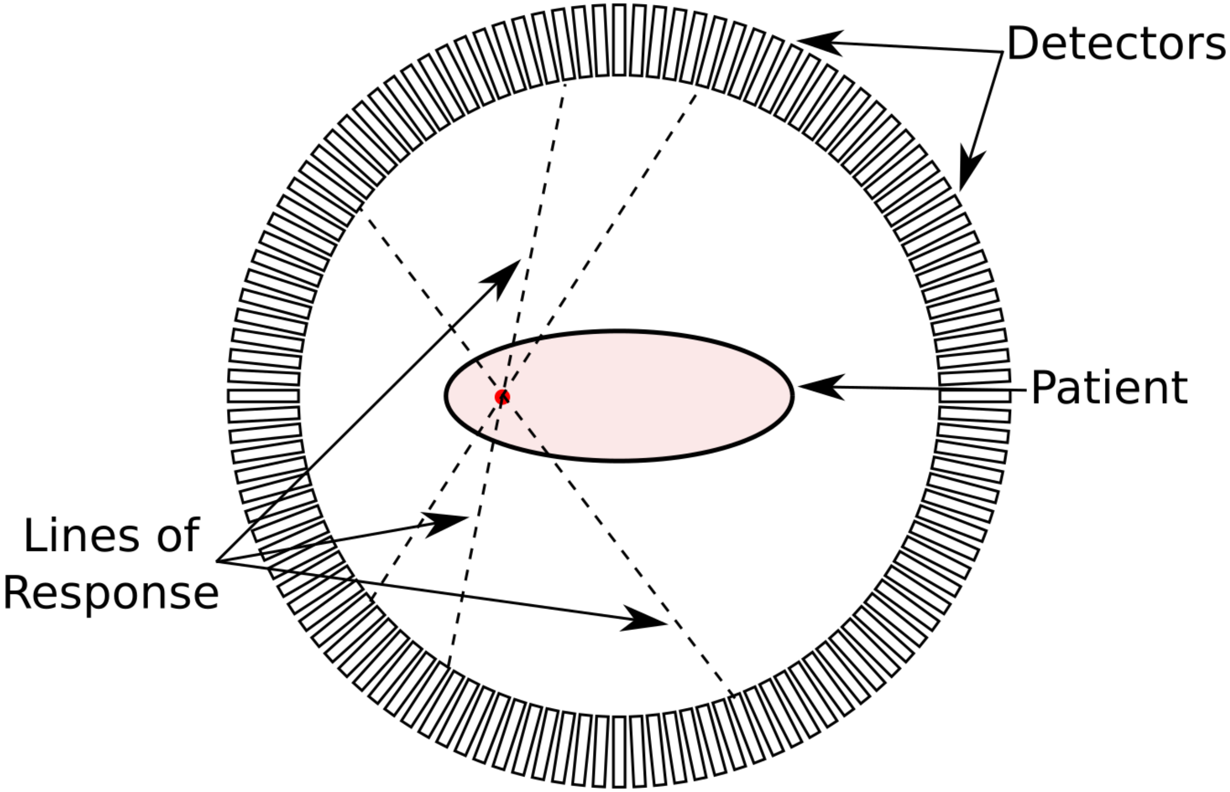
Two collinear gammas **(511 KeV)** are emitted

Positron Emission Tomography



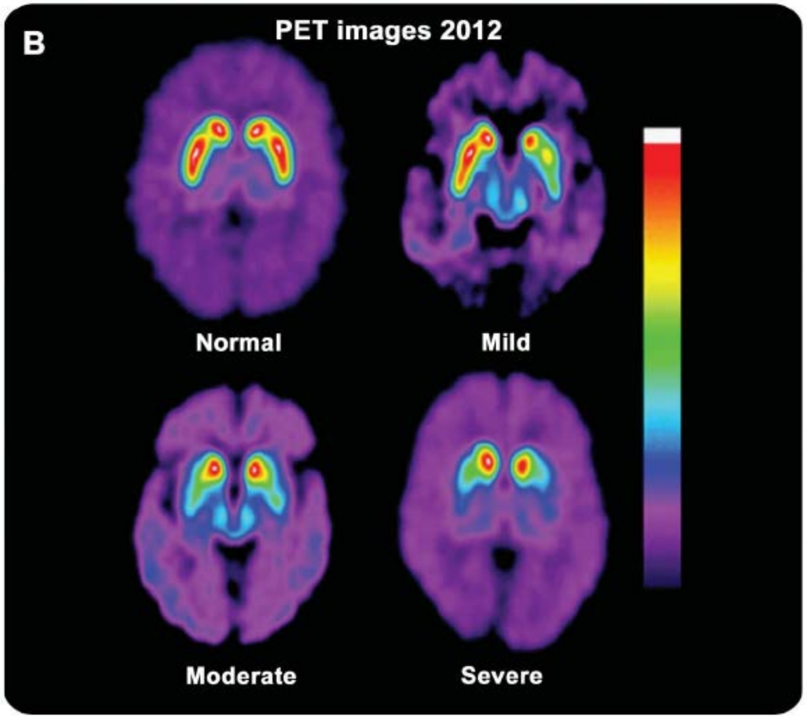
Detection of two gammas in coincidence

Positron Emission Tomography



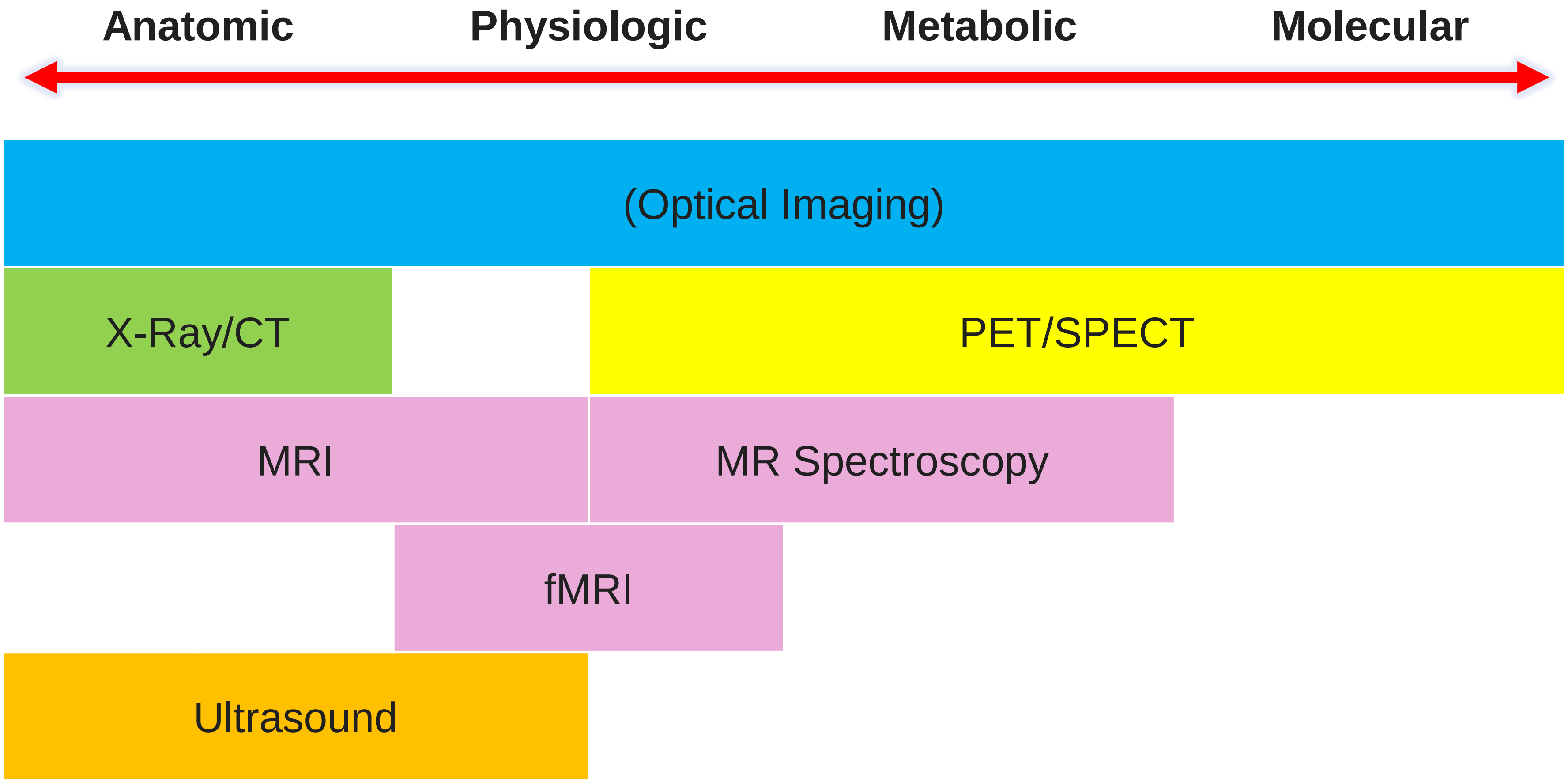
Find lines of response (LORs), intersection at source

Positron Emission Tomography



Tomographic reconstruction to get biomolecule distribution in the body

Imaging *in vivo*



Anatomic **Physiologic** **Metabolic** **Molecular**

(Optical Imaging)

X-Ray/CT

PET/SPECT

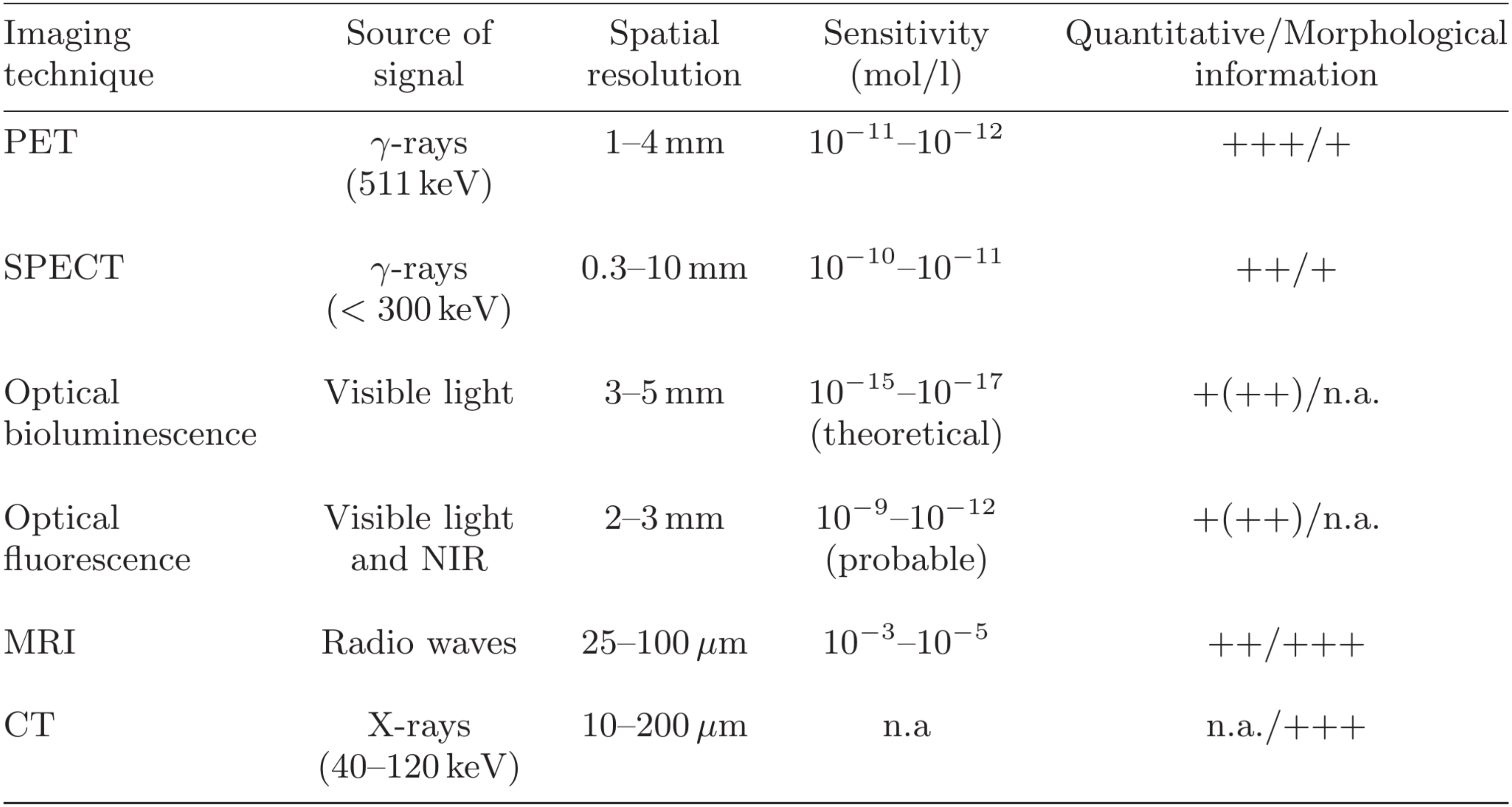
MRI

MR Spectroscopy

fMRI

Ultrasound

Why PET?



Imaging technique	Source of signal	Spatial resolution	Sensitivity (mol/l)	Quantitative/Morphological information
PET	γ-rays (511 keV)	1–4 mm	10^{-11}–10^{-12}	+++/+
SPECT	γ-rays ($<$ 300 keV)	0.3–10 mm	10^{-10}–10^{-11}	++/+
Optical bioluminescence	Visible light	3–5 mm	10^{-15}–10^{-17} (theoretical)	+(++)/n.a.
Optical fluorescence	Visible light and NIR	2–3 mm	10^{-9}–10^{-12} (probable)	+(++)/n.a.
MRI	Radio waves	25–100 μm	10^{-3}–10^{-5}	++/+++
CT	X-rays (40–120 keV)	10–200 μm	n.a	n.a./+++

from *A. Del Guerra et al., 2016 Positron Emission Tomography: Its 65 years Riv. Nuovo Cimento 39 155*

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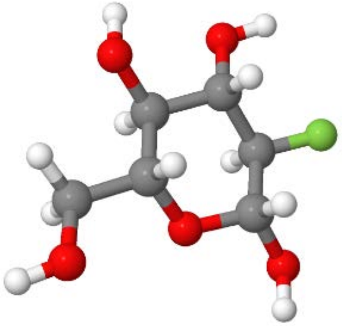
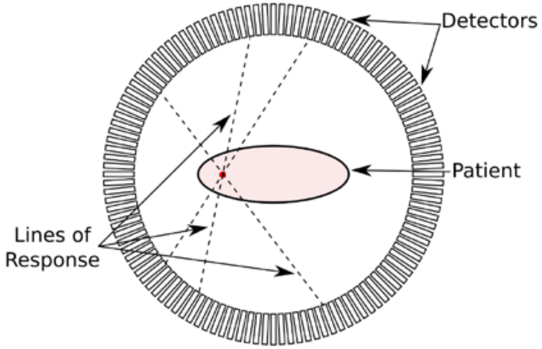
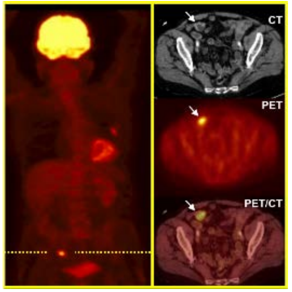
The PET system



Biology

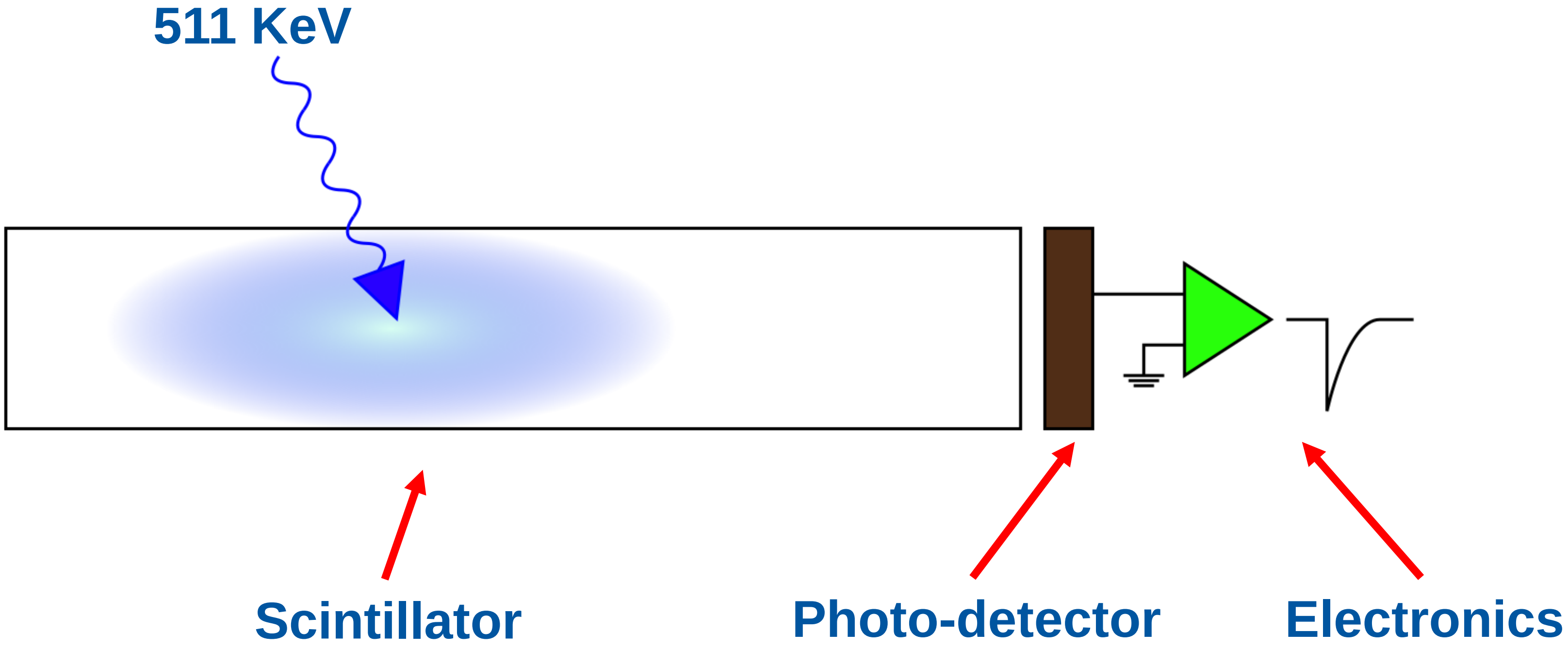


Detector



Software

Detecting radiation



From one gamma to an electrical signal

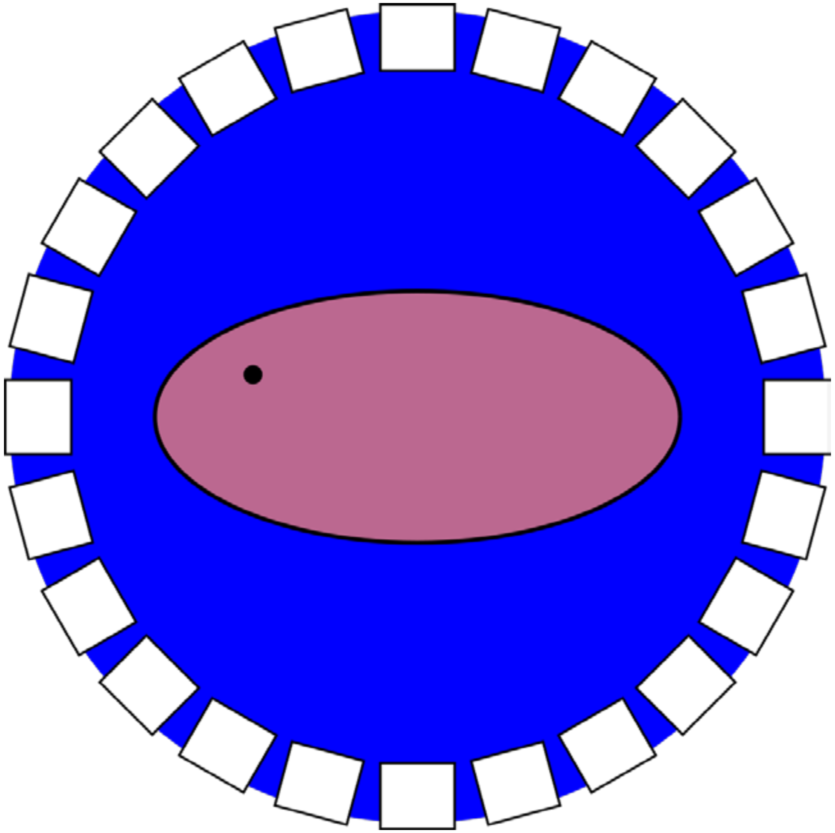
Detector requirements

- Obtain as many counts as possible
  - **high sensitivity**

- Localize counts as accurately as possible
  - **high spatial resolution**
  - **high temporal resolution**

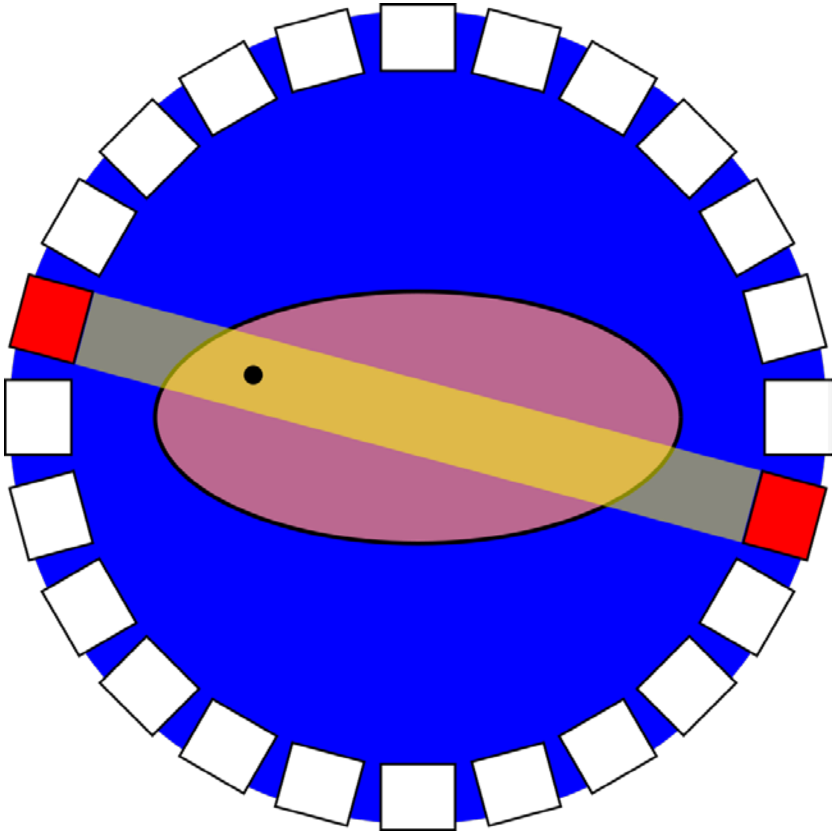
Geometry

- Cylindrical coverage and thick scintillators maximize **sensitivity**



Geometry

- Cylindrical coverage and thick scintillators maximize **sensitivity**
- **Spatial resolution** is defined by crystal section



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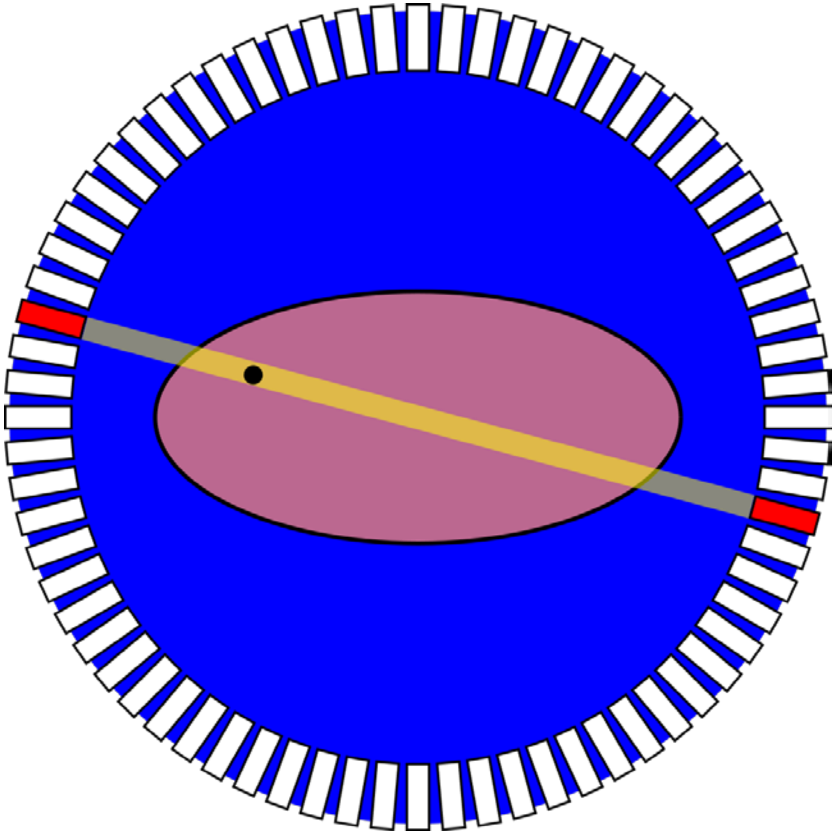

Geometry

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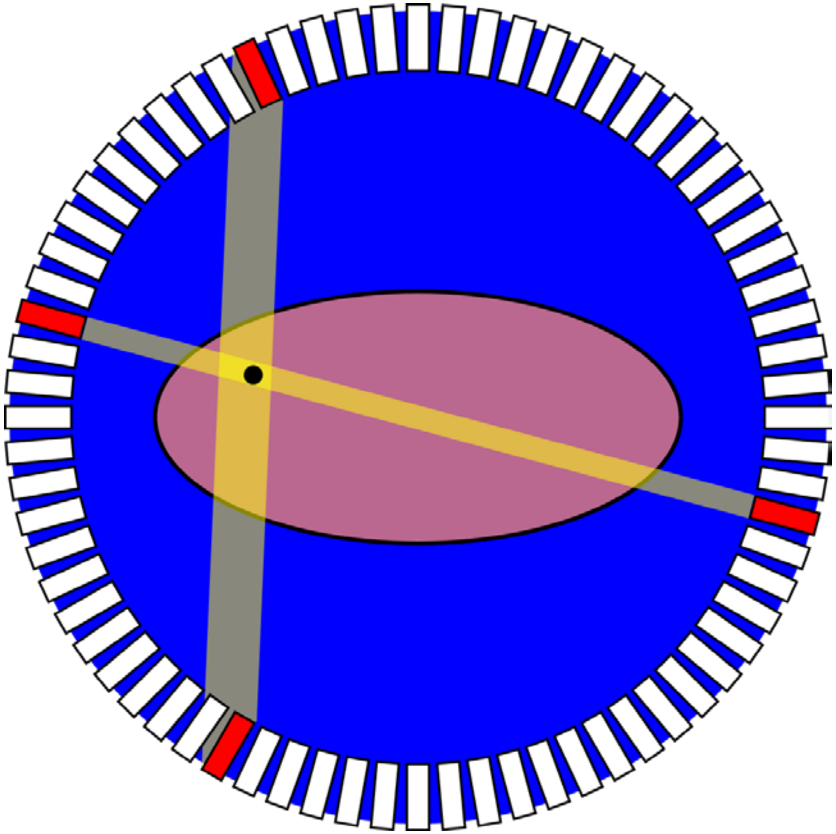
- Segmentation improves spatial resolution…
- … but at the same time resolution degrades due to parallax effect



Geometry

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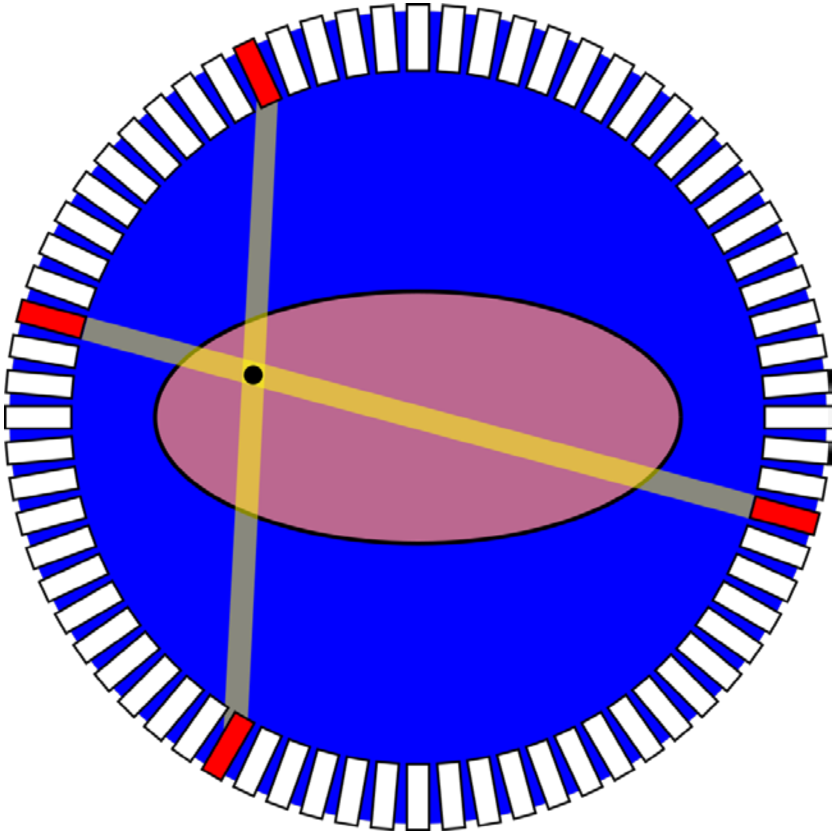
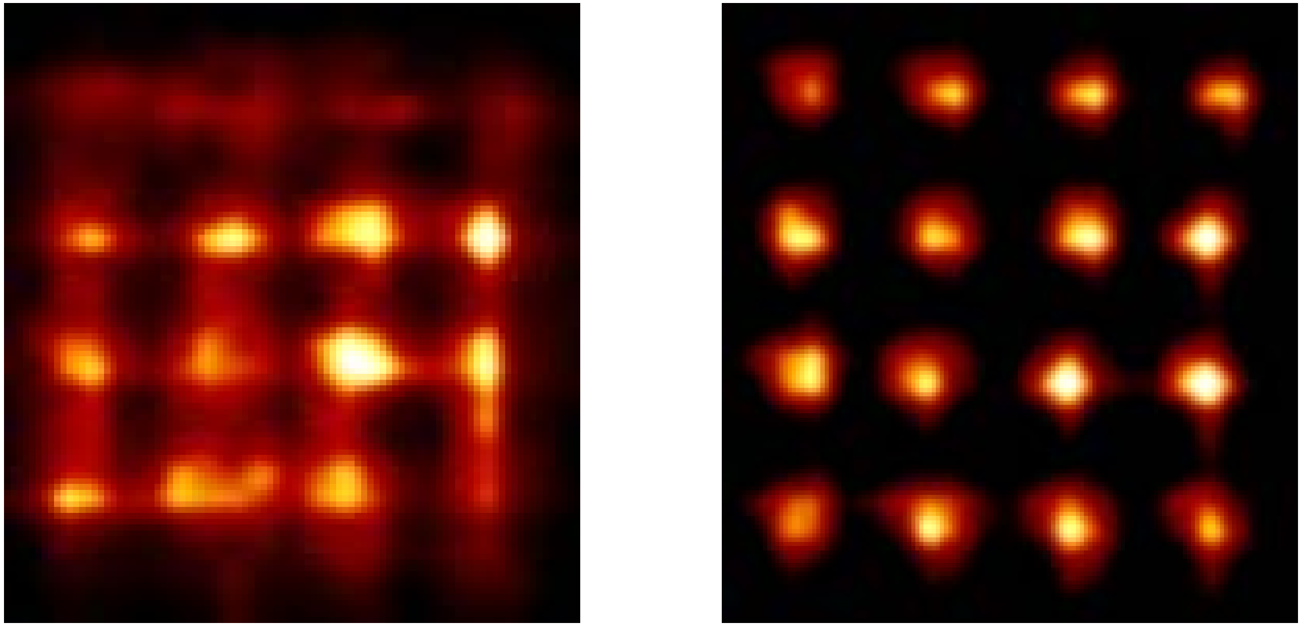
- Segmentation improves spatial resolution…
- … but at the same time resolution degrades due to parallax effect
- This can be recovered if the **Depth of Interaction** (DOI) of the gamma is measured



Depth of Interaction



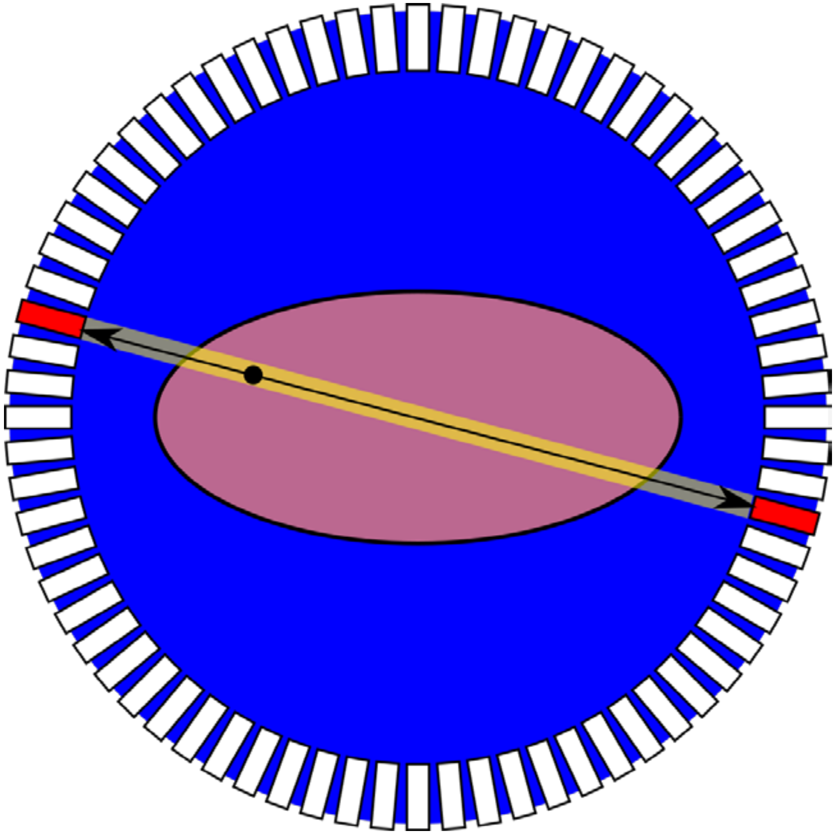
Without DOI

With DOI

^{22}Na source moved on a grid of 5 mm pitch

Coincidence sorting

- A **true coincidence** is identified if two detected gammas have:
  - energy in the 511 KeV energy window
  - difference in time of arrival within the coincidence window



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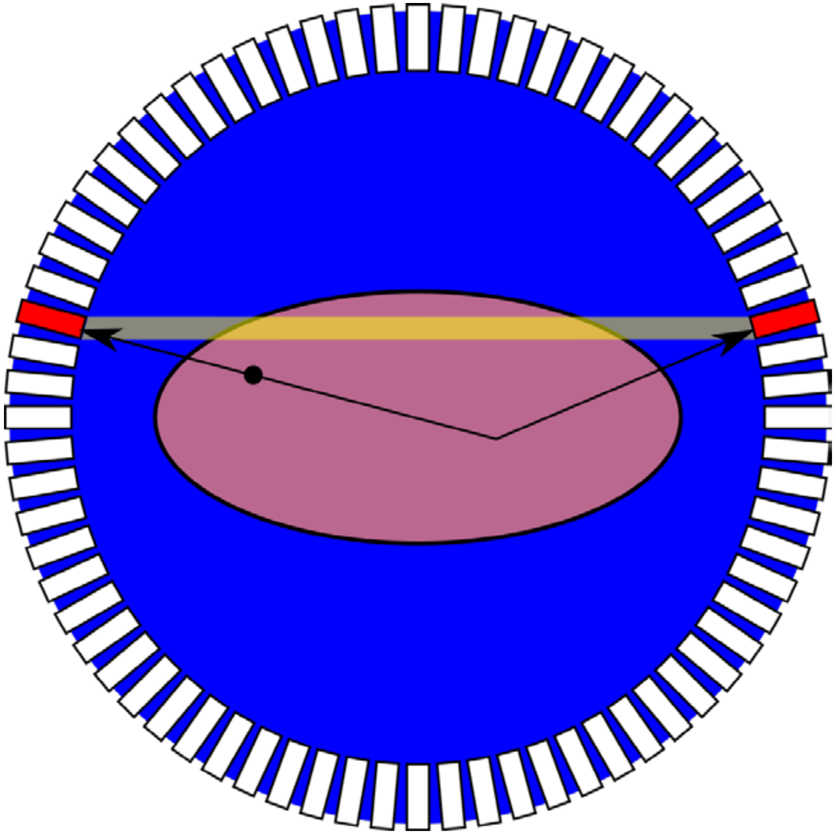
  - Good **energy resolution** is needed to reject these events (usually 10%-15% FWHM)



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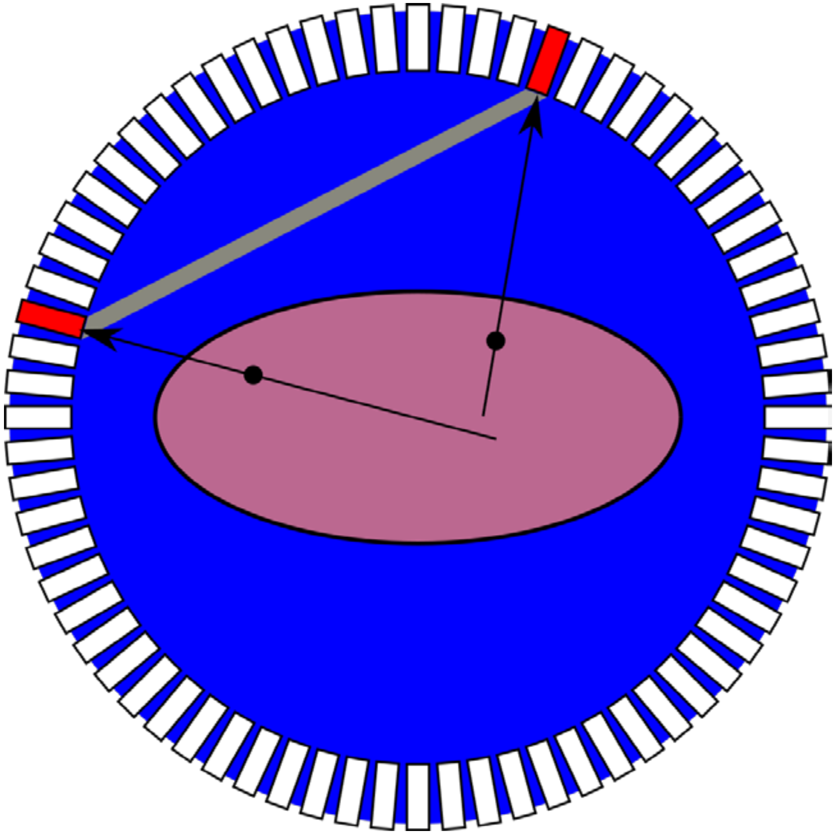
- The coincidence can be assigned to two gammas that are coming from different annihilation events and fall **randomly** within the coincidence window
  - Good **timing resolution** is needed to minimize the probability of these false coincidences (<1ns)

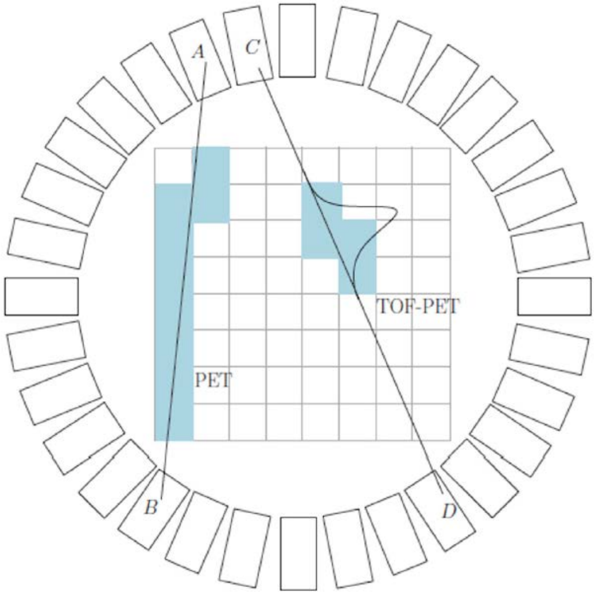


Time of Flight (TOF)

Compute the **difference in time of arrival** of gammas:

- Improve event localization along the LORs

$$\Delta x = c\frac{\Delta t}{2}$$



S. Surti, J.S. Karp - Physica Medica 32 (2016) 12–22

Time of Flight (TOF)

Compute the **difference in time of arrival** of gammas:

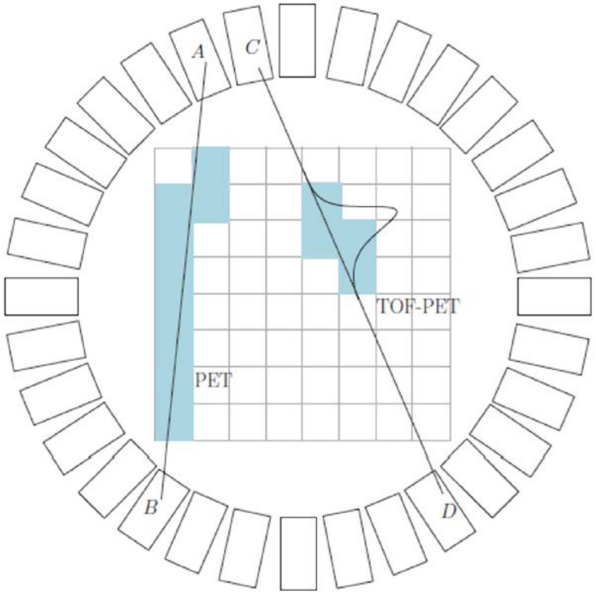
- Improve event localization along the LORs

$$\Delta x = c\frac{\Delta t}{2}$$

- Decrease noise correlation in overlapping LORs, improve signal-to-noise ratio (SNR)

$$SNR_{TOF} \sim \sqrt{\frac{D}{\Delta x}} \cdot SNR_{CONV}$$

D = effective object diameter



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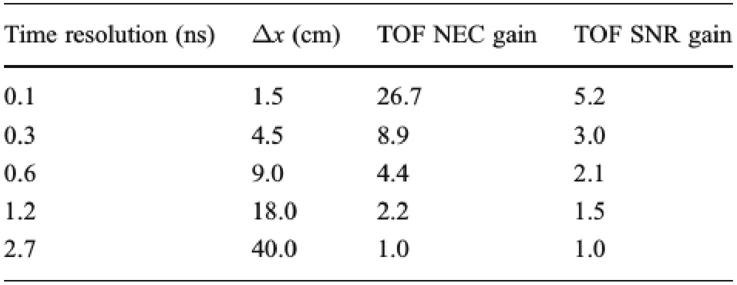

Time resolution (ns)	Δx (cm)	TOF NEC gain	TOF SNR gain
0.1	1.5	26.7	5.2
0.3	4.5	8.9	3.0
0.6	9.0	4.4	2.1
1.2	18.0	2.2	1.5
2.7	40.0	1.0	1.0

M. Conti - Eur J Nucl Med Mol Imaging (2011) 38:1147–1157

Benefits of TOF

- Improved **lesion detectability** while keeping scanning time constant

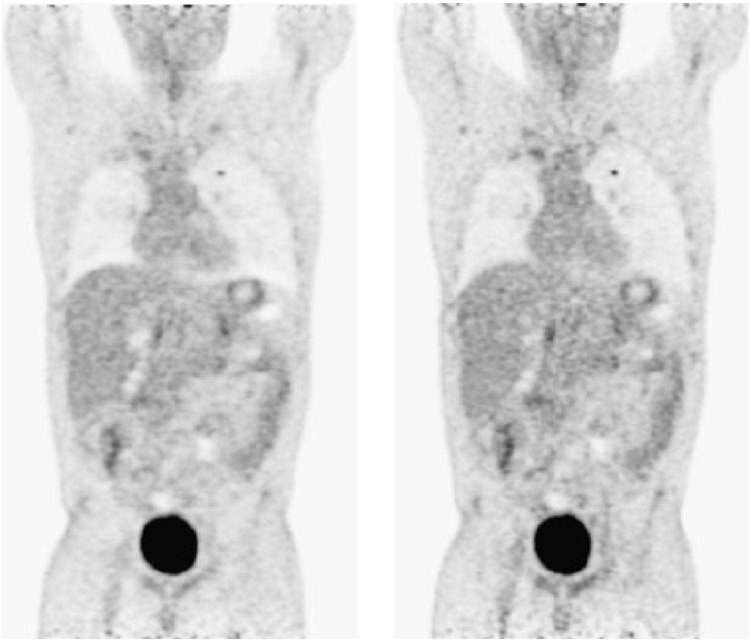

Fig. 1 Coronal images reconstructed from a non-TOF scan (*left*) and a TOF scan (*right*) in a patient with lung cancer. The acquisition time was 3 min per bed position for both images. At the same number of counts, the image quality is better with the TOF reconstruction

M. Conti - Eur J Nucl Med Mol Imaging (2011) 38: 1147

Benefits of TOF

- Improved **lesion detectability** while keeping scanning time constant
- Reduced **scan times** for the same lesion detectability

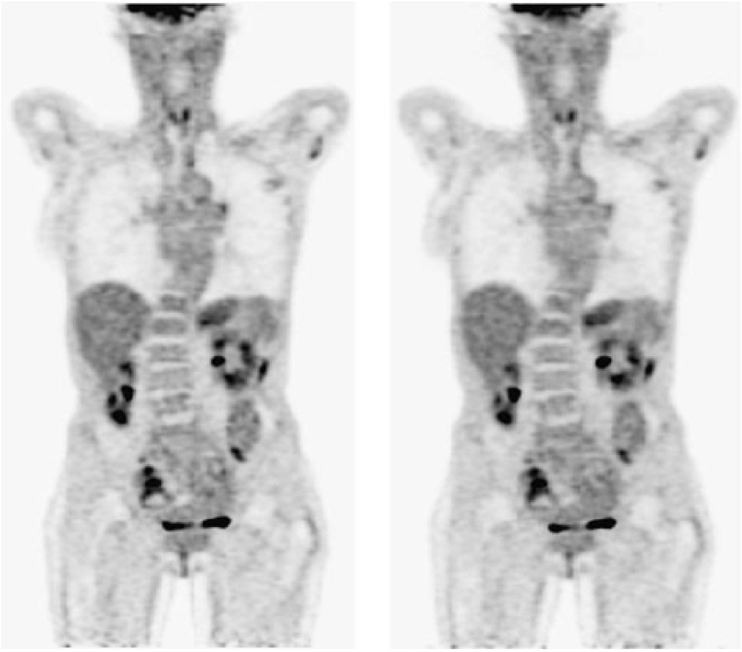

Fig. 2 Coronal images reconstructed from a non-TOF scan (*left*) and a TOF scan (*right*). The acquisition time was 2 min per bed position for the non-TOF scan and 1 min per bed position for the TOF scan. The quality of the non-TOF image and that of the TOF image with half of the counts are similar

M. Conti - Eur J Nucl Med Mol Imaging (2011) 38: 1147

Benefits of TOF

- Improved **lesion detectability** while keeping scanning time constant
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- **Fewer iterations** of reconstruction algorithms required to maximize lesion contrast -> lower image noise

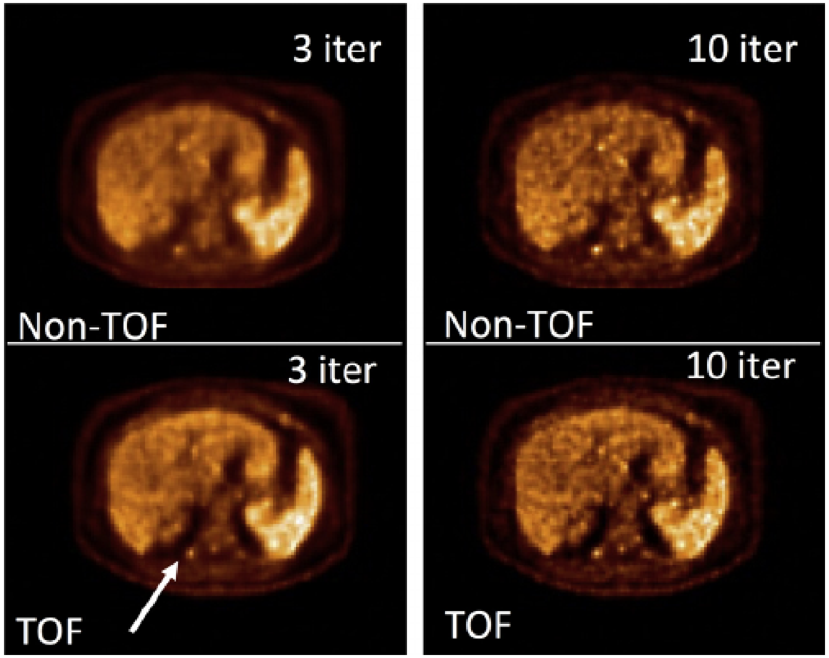

Figure 2. Reconstructed transverse slices of a clinical ^{18}F-FDG study. As indicated, images are shown for Non-TOF and TOF reconstruction and for iterations 3 and 10 of the reconstruction algorithm. The arrow indicates the lesion for which an accurate SUV is measured after 3 iterations of the TOF reconstruction algorithm.

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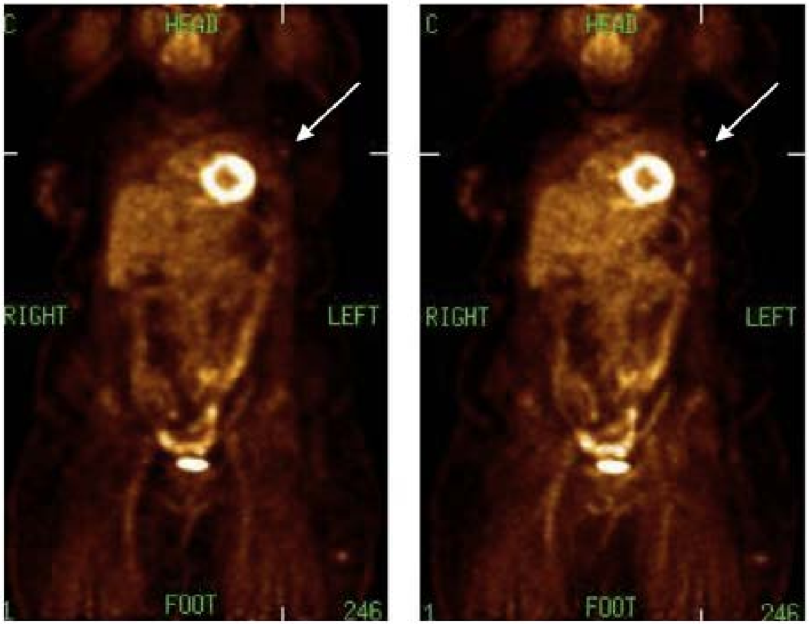
- Better lesion detectability for **larger objects**



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- **Fewer iterations** of reconstruction algorithms required to maximize lesion contrast -> lower image noise
- Better lesion detectability for **larger objects**
- Better image reconstruction for **limited angle** PET acquisitions

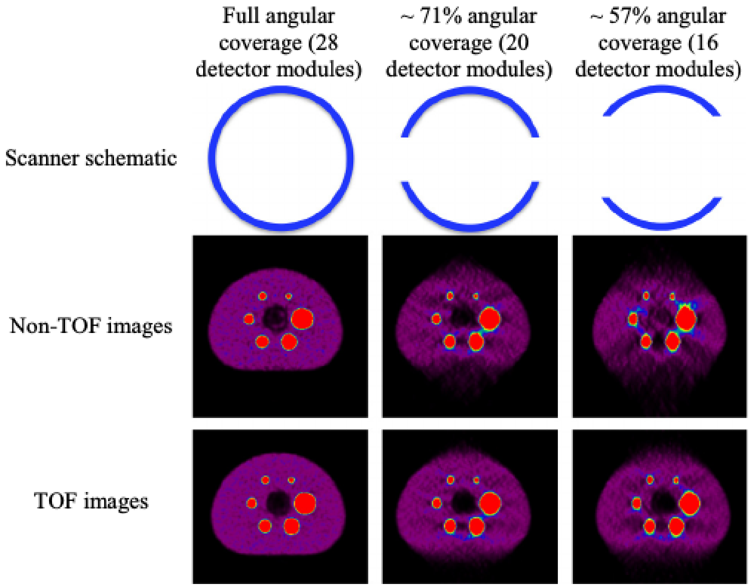

Figure 4. Reconstructed images from a NEMA image quality phantom using full or partial angular data acquired on a clinical TOF PET/CT. The six hot spheres in a ring have diameters of 37, 28, 22, 17, 13, and 10 mm and have an activity uptake of 9.7:1 with respect to background. The central cold region is a lung insert.

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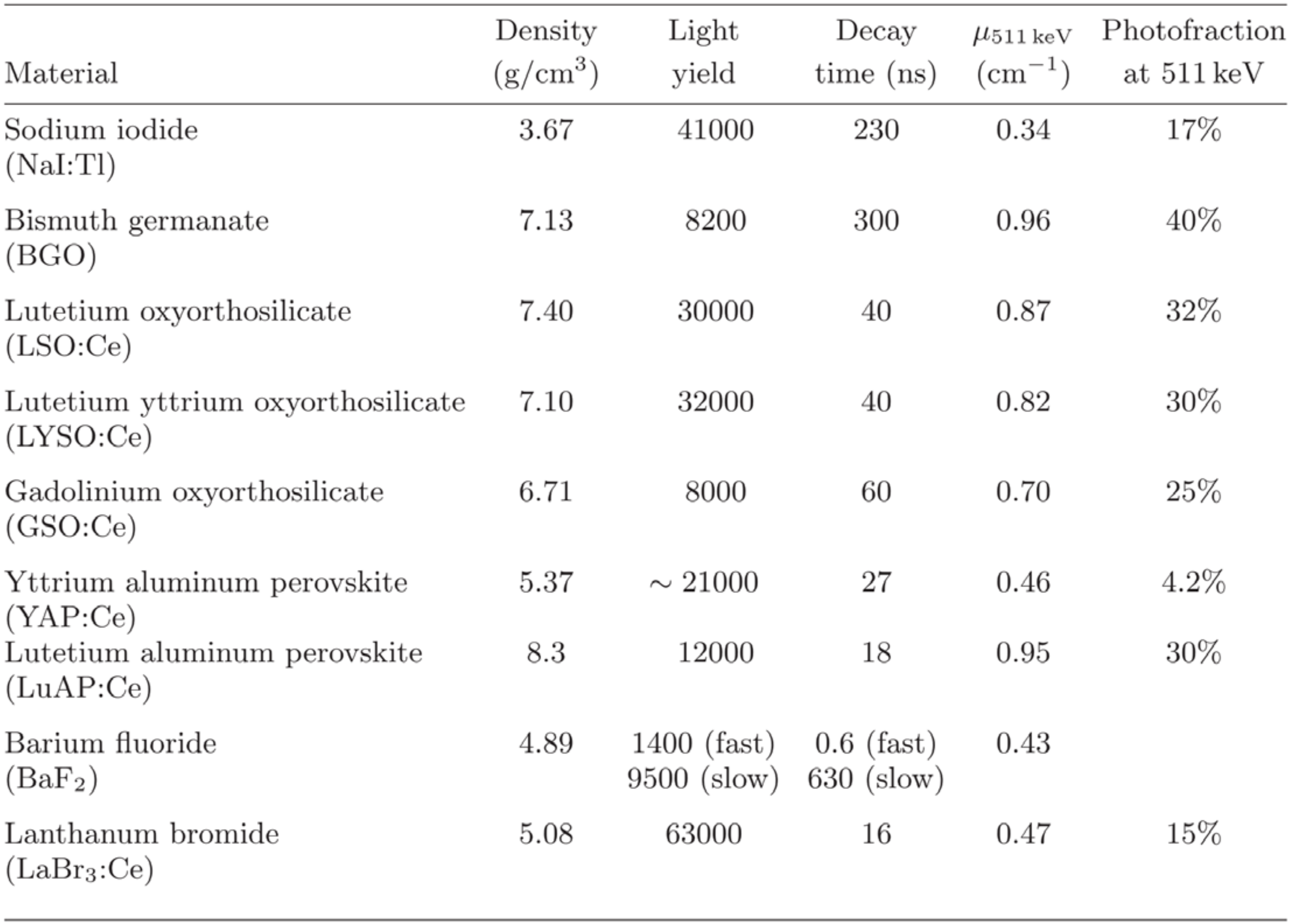
Scintillator requirements

Requirement	Parameter
Stopping power	Z_{eff}
Spatial resolution	Photofraction
Energy resolution	Light output Homogeneity Proportionality
Timing resolution	Light output Timing profile

Common PET scintillators

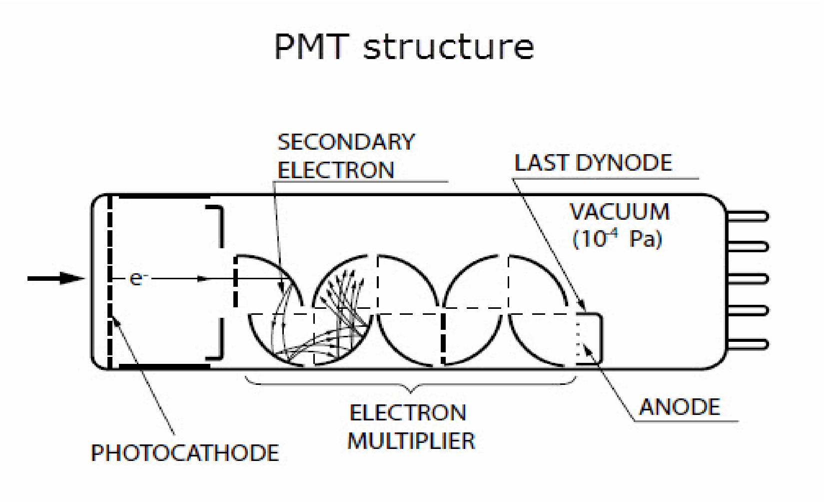


Material	Density ($\mathrm{g/cm^3}$)	Light yield	Decay time (ns)	$\mu_{511\,\mathrm{keV}}$ ($\mathrm{cm^{-1}}$)	Photofraction at 511 keV
Sodium iodide (NaI:Tl)	3.67	41000	230	0.34	17%
Bismuth germanate (BGO)	7.13	8200	300	0.96	40%
Lutetium oxyorthosilicate (LSO:Ce)	7.40	30000	40	0.87	32%
Lutetium yttrium oxyorthosilicate (LYSO:Ce)	7.10	32000	40	0.82	30%
Gadolinium oxyorthosilicate (GSO:Ce)	6.71	8000	60	0.70	25%
Yttrium aluminum perovskite (YAP:Ce)	5.37	~ 21000	27	0.46	4.2%
Lutetium aluminum perovskite (LuAP:Ce)	8.3	12000	18	0.95	30%
Barium fluoride ($\mathrm{BaF_2}$)	4.89	1400 (fast) 9500 (slow)	0.6 (fast) 630 (slow)	0.43	
Lanthanum bromide ($\mathrm{LaBr_3}$:Ce)	5.08	63000	16	0.47	15%

from *A. Del Guerra et al., 2016 Positron Emission Tomography: Its 65 years Riv. Nuovo Cimento 39 155*

Photodetectors

- **Photomultiplier Tubes (PMTs):**
  - Very high gain (up to 10^9)
  - Operated at high voltage
  - Relatively bulky and expensive
  - Sensitive to magnetic fields



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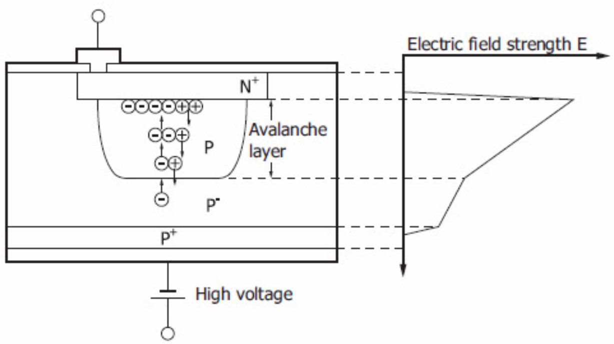
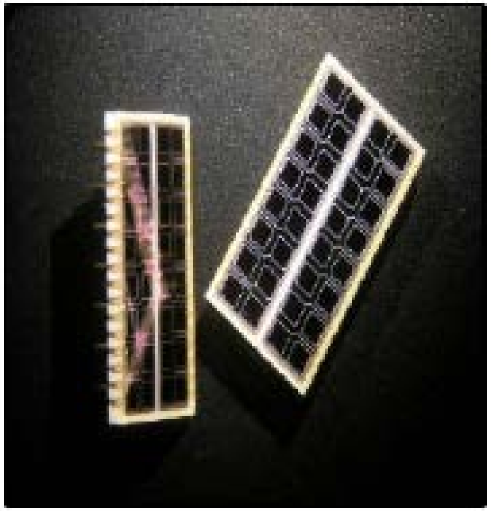
- **Avalanche Photodiodes (APDs):**
  - Relatively low gain (10^2-10^3)
  - Operated at low voltage
  - Compact, 1-to-1 coupling
  - Insensitive to magnetic fields

APD structure





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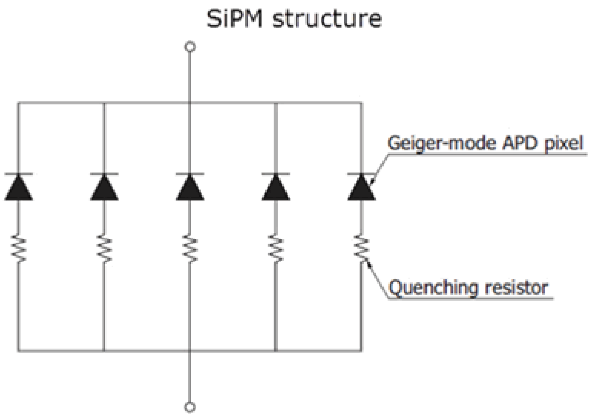
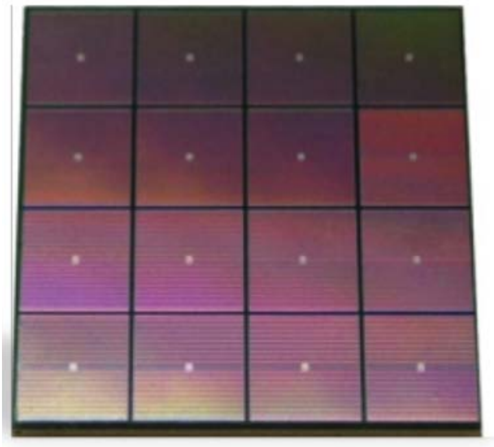
  - Compact, 1-to-1 coupling
  - Insensitive to magnetic fields
- **Silicon Photomultipliers (SiPMs):**
  - High gain (up to 10^7)
  - Operated at low voltage
  - Compact
  - Insensitive to magnetic fields
  - Better timing performance





Spatial resolution

$$\Delta x = 1.25\sqrt{(d/2)^2 + b^2 + (0.0022D)^2 + r^2 + p^2}$$

W.W. Moses, S.E. Derenzo, J. Nucl. Med. 34 (1993) 101P

Spatial resolution

$$\Delta x = 1.25\sqrt{\overbrace{(d/2)^2}^{\textbf{Crystal size}}+b^2+(0.0022D)^2+r^2+\overbrace{p^2}^{\textbf{Parallax error}}}$$

W.W. Moses, S.E. Derenzo, J. Nucl. Med. 34 (1993) 101P

Spatial resolution

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W.W. Moses, S.E. Derenzo, J. Nucl. Med. 34 (1993) 101P

Spatial resolution

$$\Delta x = \underbrace{1.25}_{\textbf{Reconstruction}}\sqrt{\overbrace{(d/2)^2}^{\textbf{Crystal size}}+\overbrace{b^2}^{\textbf{Positioning precision}} + (0.0022D)^2+r^2 + \overbrace{p^2}^{\textbf{Parallax error}}}$$

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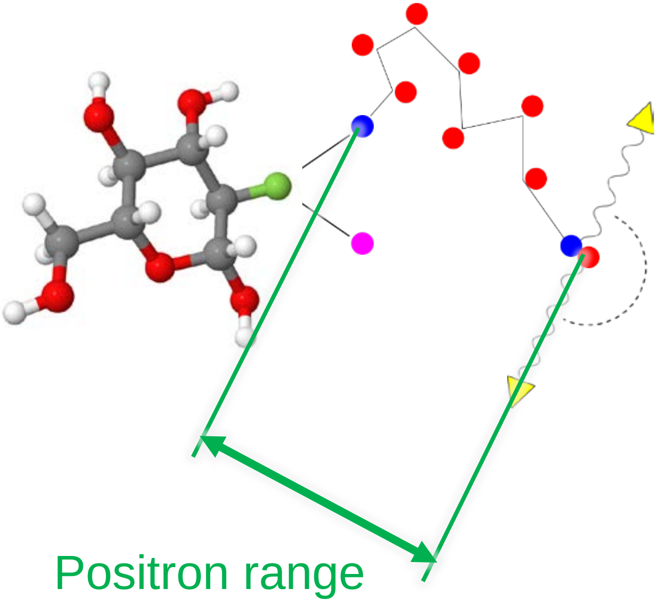
W.W. Moses, S.E. Derenzo, J. Nucl. Med. 34 (1993) 101P

Positron range

- The two 511 KeV gammas are not emitted at the site of positron emission
- Positron loses energy by **scattering** in the tissue undergoing a random walk
- The range is determined by the **initial energy** of the positron



PET signal is blurred “at the source”

Common β^+ emitters

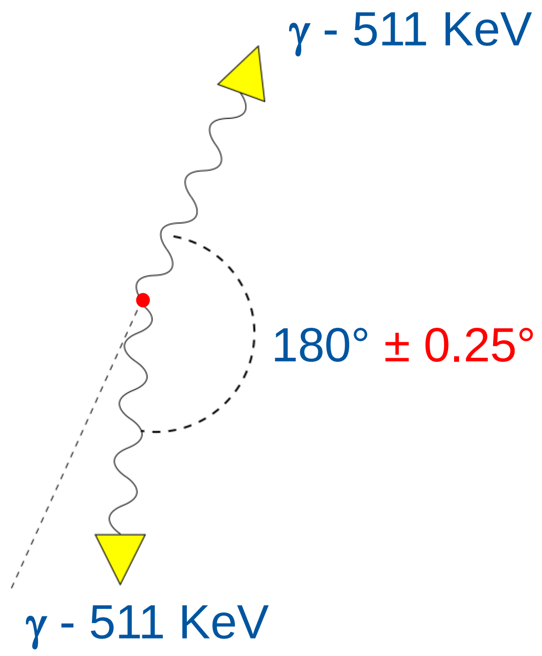
Radioisotope	Half-life (min)	Positron average kinetic energy (MeV)	Positron kinetic energy endpoint (MeV)	Positron average range in water (mm)
^{11}C	20.4	0.385	0.960	1.2
^{13}N	10.0	0.491	1.198	1.6
^{15}O	2.0	0.735	1.732	2.8
^{18}F	109.8	0.242	0.633	0.6

from *A. Del Guerra et al., 2016 Positron Emission Tomography: Its 65 years Riv. Nuovo Cimento 39 155*

- Several biomolecules can be labeled
  - sugar (e.g. FDG), enzymes, antibodies…
- But keep in mind 2 parameters:
  - **Half life** (production on site, activity decay during exam)
  - **Positron range** (signal blurring)

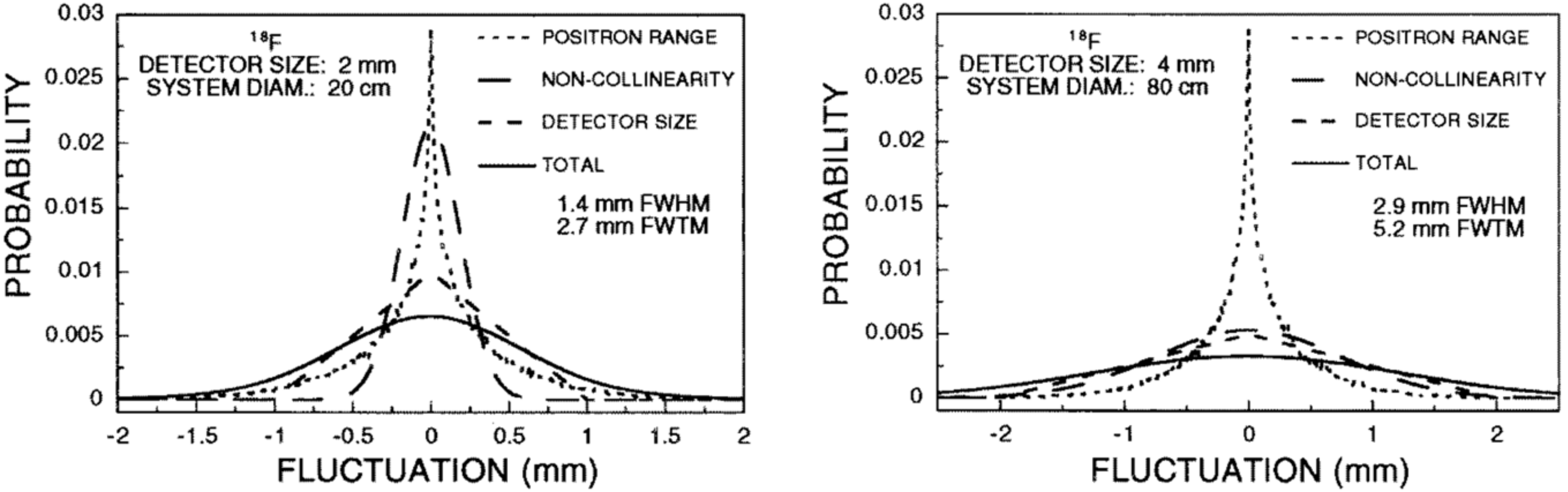
Collinearity

- Bound energy of the electron annihilating with the positron is **not negligible**
- The annihilation photons are emitted with angle **180° ± 0.25°**
- Further **blurring** on the final image, depending on scanner diameter D **(-> 0.0022D)**



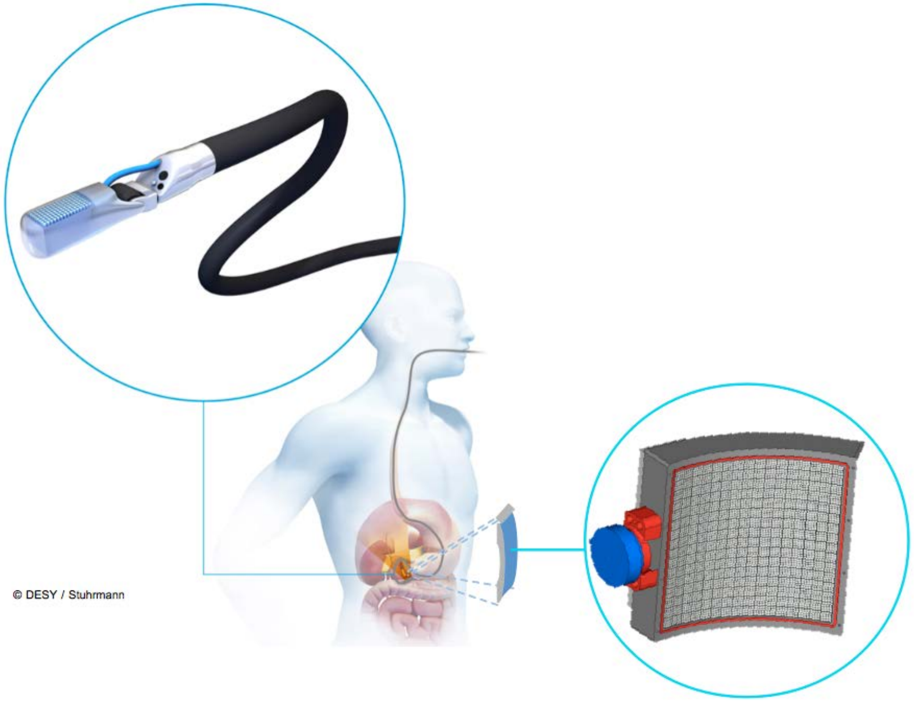
PET signal is blurred “at the source”

Resolution limit



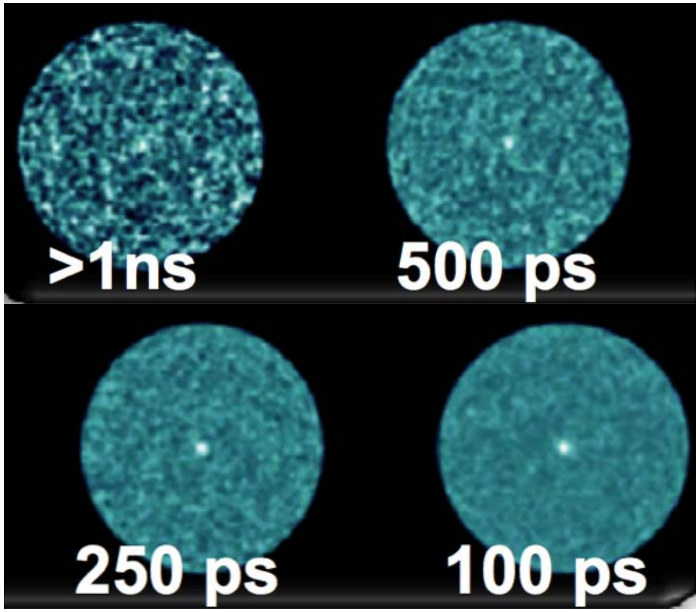
C.S. Levin and E. J. Hoffman, Phys. Med. Biol. 44 (1999) 781–799

Time of flight: why more?


© DESY / Stuhrmann

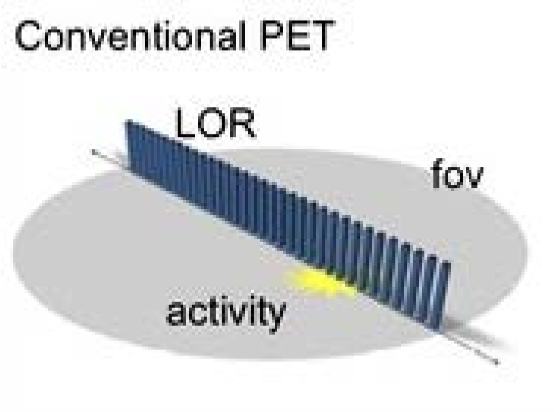
- **@200ps** CTR -> Better background rejection for **small organs** (e.g. EndoTOFPET)

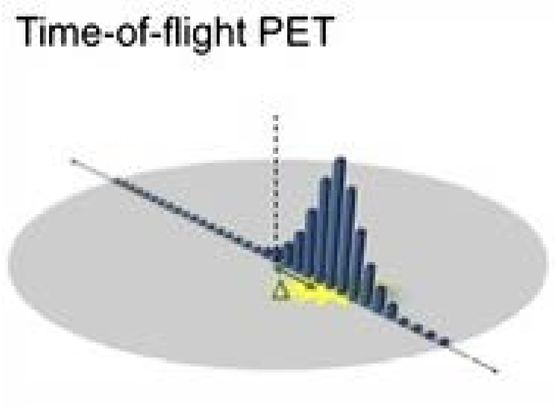
Time of flight: why more?



- **@200ps** CTR -> Better background rejection for **small organs** (e.g. EndoTOFPET)
- **@100ps** CTR -> **SNR** improved by factor 5 (potential sensitivity gain x25)

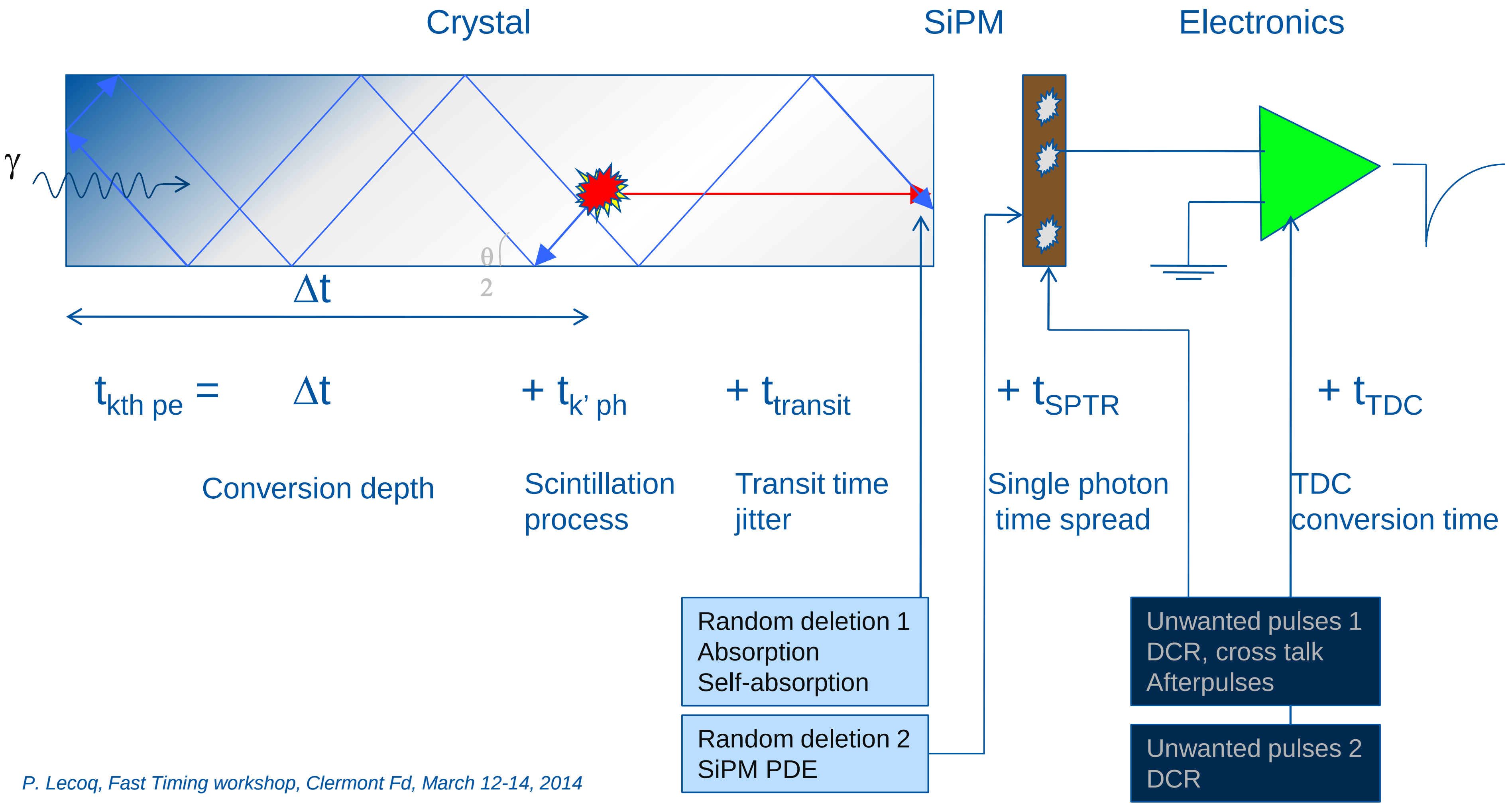
Time of flight: why more?





- **@200ps** CTR -> Better background rejection for **small organs** (e.g. EndoTOFPET)
- **@100ps** CTR -> **SNR** improved by factor 5 (potential sensitivity gain x25)
- **@ 10ps** CTR -> Direct 3D information, **reconstructionless PET** and **online image**

The detection chain



Crystal

SiPM

Electronics

$$t_{kth\ pe} = \Delta t + t_{k'\ ph} + t_{transit} + t_{SPTR} + t_{TDC}$$

Conversion depth

Scintillation process

Transit time jitter

Single photon time spread

TDC conversion time

Random deletion 1
Absorption
Self-absorption

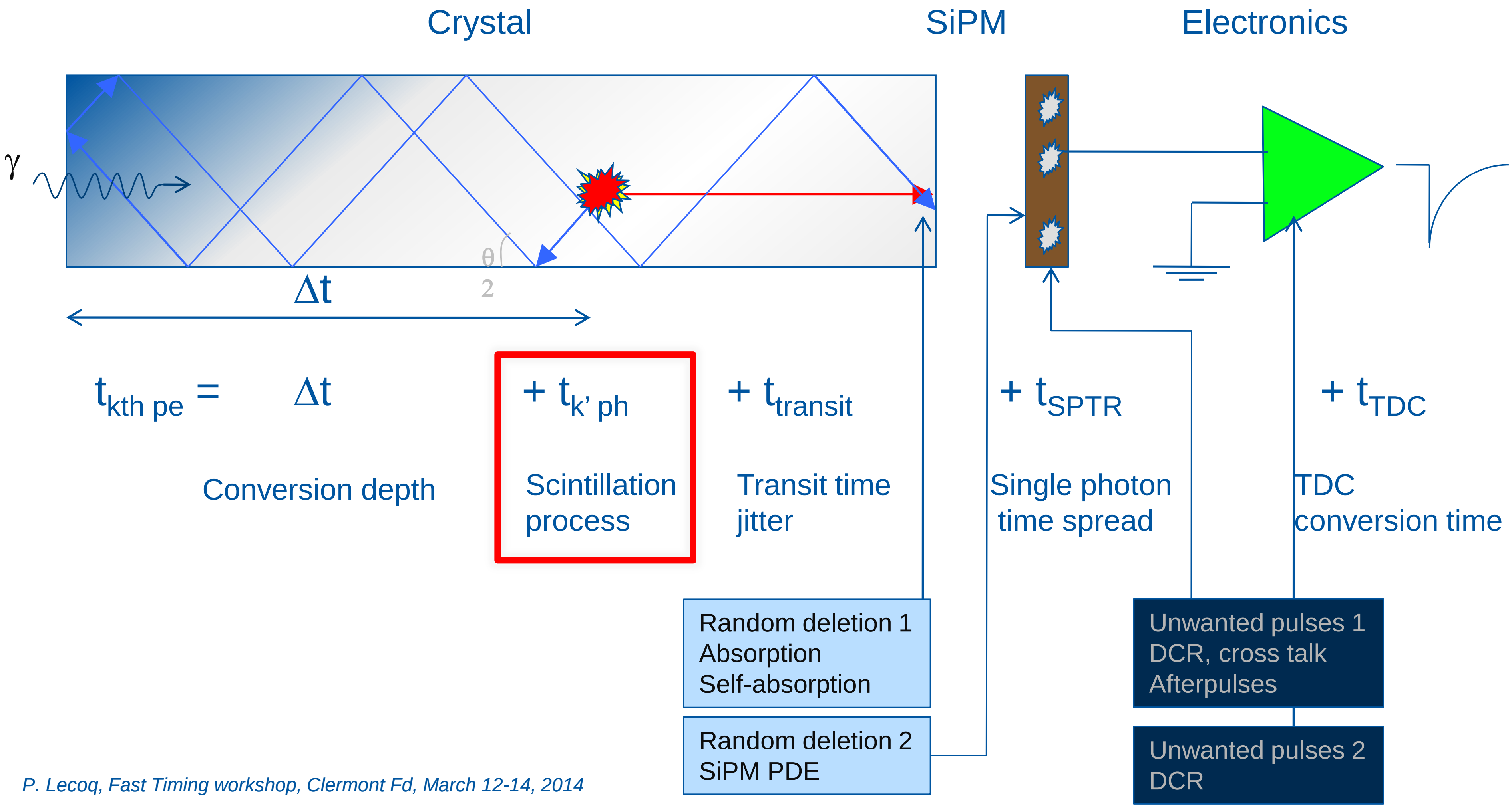
Random deletion 2
SiPM PDE

Unwanted pulses 1
DCR, cross talk
Afterpulses

Unwanted pulses 2
DCR

P. Lecoq, Fast Timing workshop, Clermont Fd, March 12-14, 2014

The detection chain



Crystal

SiPM

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P. Lecoq, Fast Timing workshop, Clermont Fd, March 12-14, 2014

Scintillators and TOF

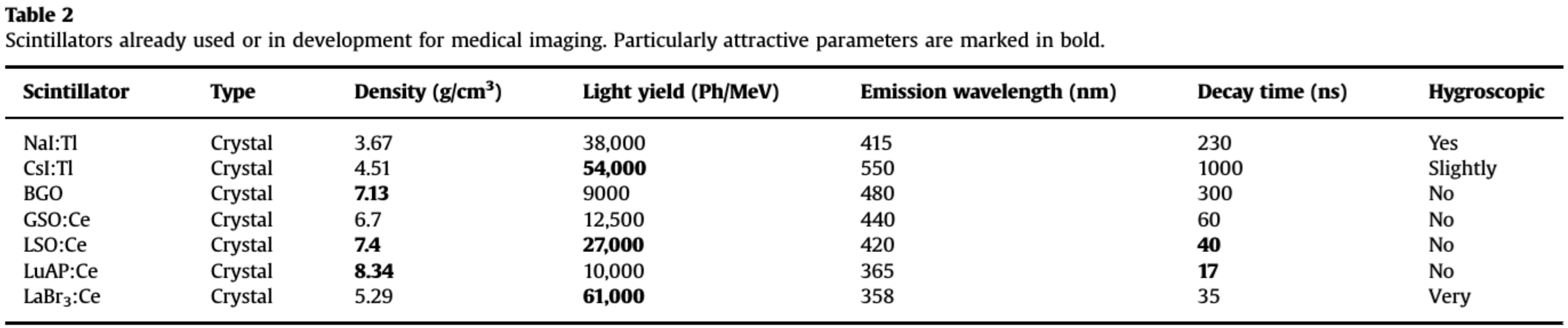


Table 2
Scintillators already used or in development for medical imaging. Particularly attractive parameters are marked in bold.

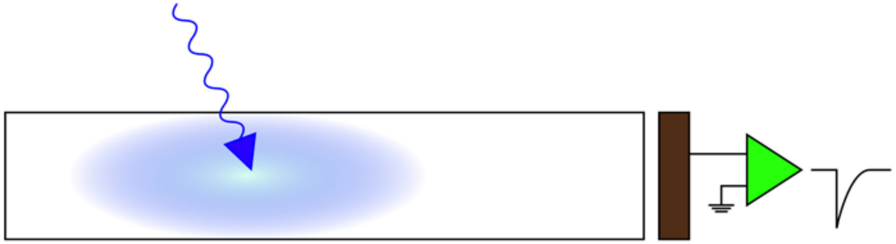
Scintillator	Type	Density (g/cm^3)	Light yield (Ph/MeV)	Emission wavelength (nm)	Decay time (ns)	Hygroscopic
NaI:Tl	Crystal	3.67	38,000	415	230	Yes
CsI:Tl	Crystal	4.51	**54,000**	550	1000	Slightly
BGO	Crystal	**7.13**	9000	480	300	No
GSO:Ce	Crystal	6.7	12,500	440	60	No
LSO:Ce	Crystal	**7.4**	**27,000**	420	**40**	No
LuAP:Ce	Crystal	**8.34**	10,000	365	**17**	No
$LaBr_3$:Ce	Crystal	5.29	**61,000**	358	35	Very

P. Lecoq, NIM A 809 (2016) 130–139

- Scintillation **intrinsically** limits CTR to 100ps (time jitter in relaxation process)

- Research ongoing on **ultrafast emission** mechanisms
  - Cerenkov photons
  - Hot Intra Band Luminescence
  - Nanocrystals

PET detector: summary



- Limiting factors to **spatial resolution**:
  - Geometrical (segmentation, size, parallax)
  - Mathematical (no direct info, need for a reconstruction)
  - Intrinsic (positron range, collinearity)

- Limiting factors for **energy resolution**
  - Scintillators
  - (Photodetectors)

- Limiting factors for **timing resolution**
  - Main long term limitation: light production mechanism

Outline

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    - ClearPET
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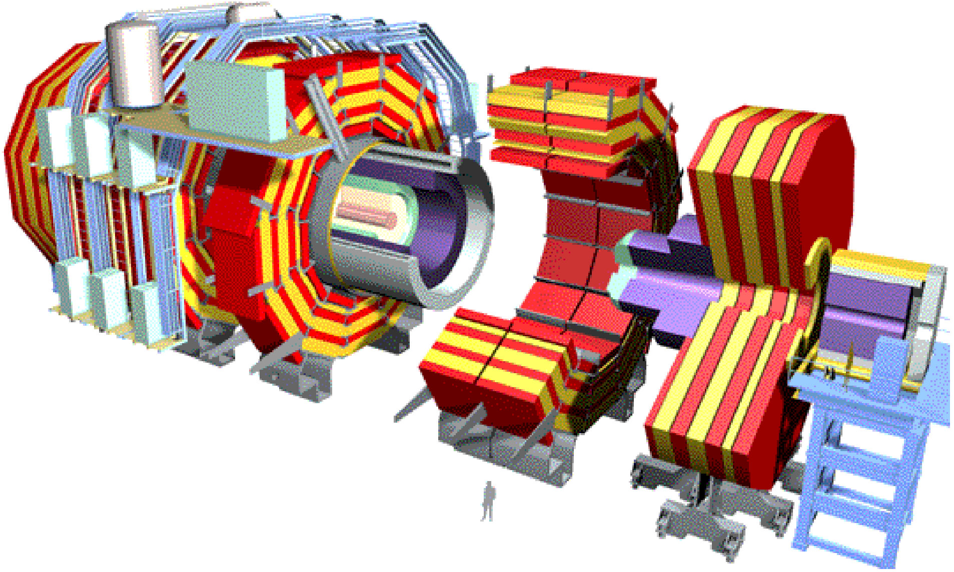
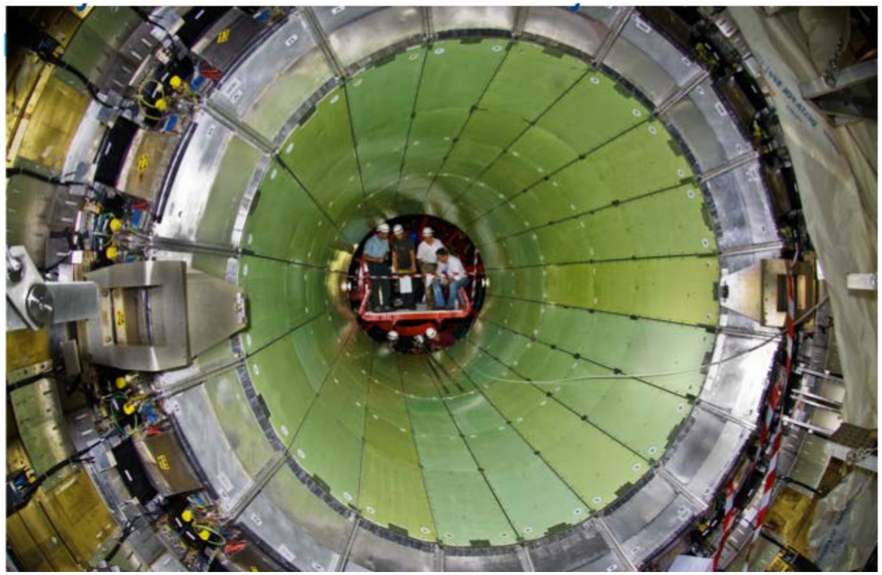
The Crystal Clear Collaboration





- International collaboration created in 1991 at CERN
- Develop scintillating materials suitable for the LHC collider
- Based on CCC work, CMS in 1994 chose $PbWO_4$ for its ECAL

The Crystal Clear Collaboration

- Today **29 institutes** all over the world
- Main activities:
  - Generic activities on inorganic scintillators and their understanding
  - Scintillation mechanisms, timing properties, radiation hardness
  - Generic activities of photo-detectors
  - Detector developments, mainly for HEP and medical imaging
- Strong focus on **fast timing** in recent years, through CCC network and European projects initiated by CERN group

ENDO TOFPET US
Endoscopic TOFPET & Ultrasound

THE PICOSEC
MC-NET PROJECT

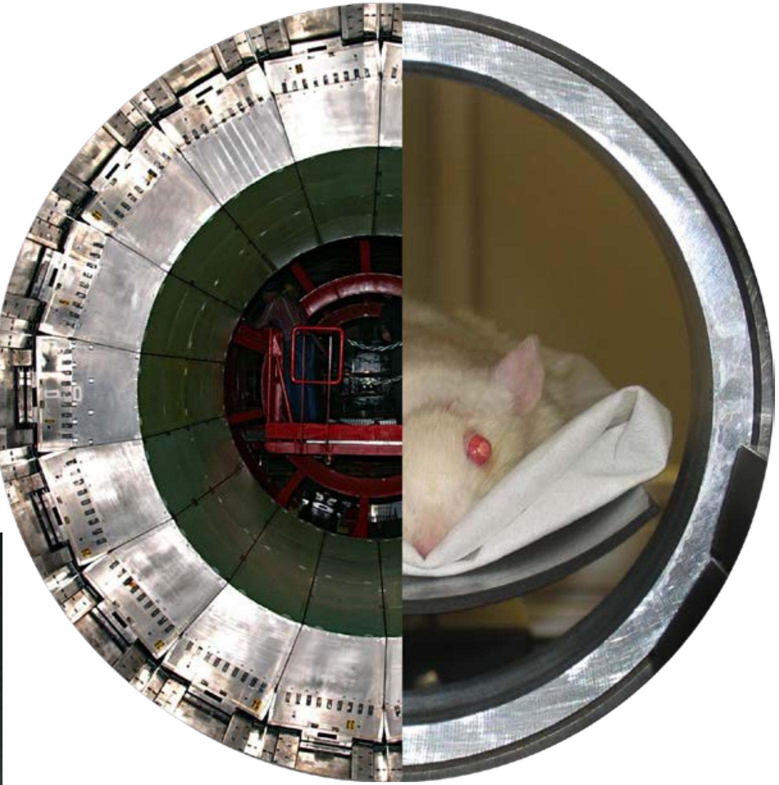
FAST
Fast Advanced Scintillator Timing

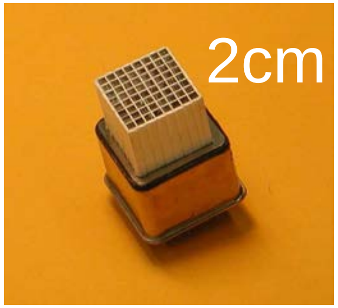
TICAL

Similar imaging techniques

CMS ECAL

PET





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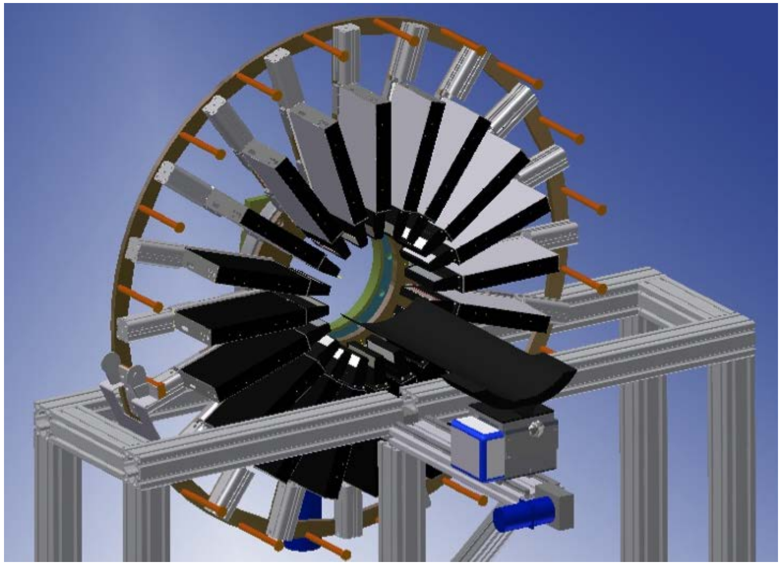
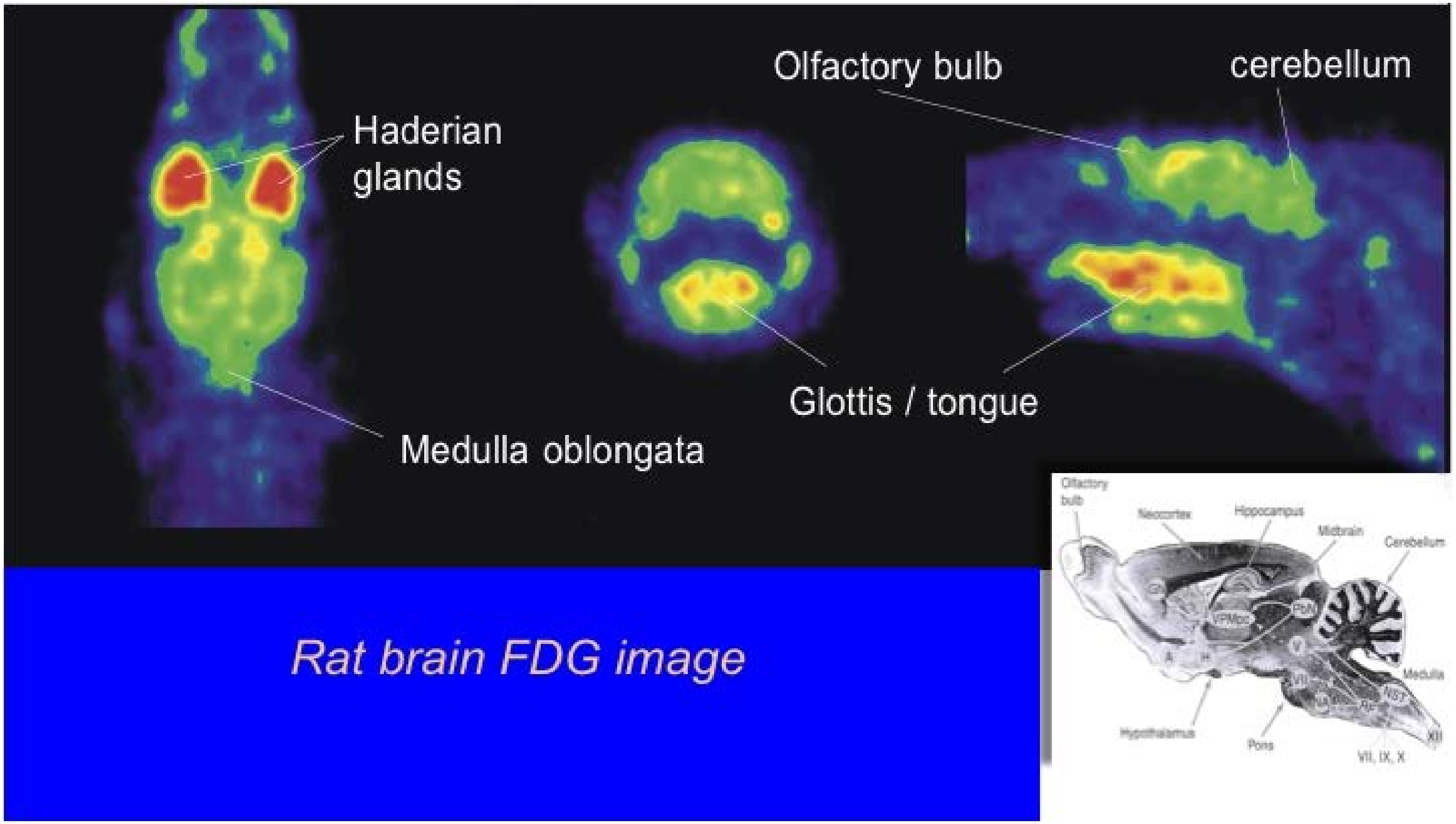
ClearPET



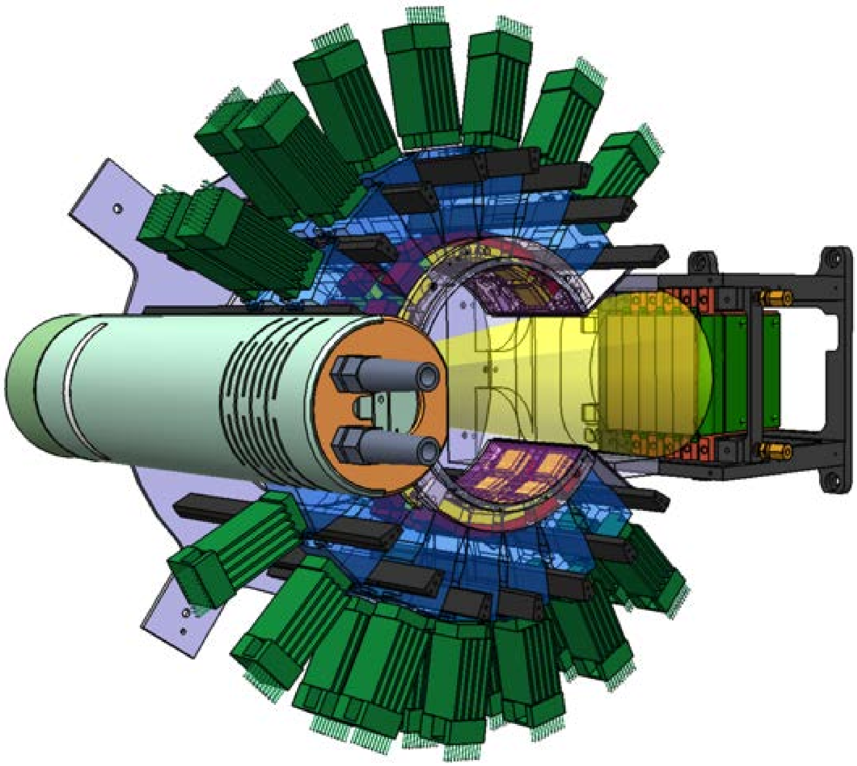
- Target: pre-clinical PET on small animals
  - Developed in 1995, 4 prototypes built
  - Futher developments ongoing in Marseille an Aachen
- Key features:
  - Dual layer LYSO/LuYAP phoswich for DOI (2x2x10mm^3 + 2x2x10mm^3)
  - 80 multi-channel PMTs
  - Spatial resolution 1.5 mm in center of FOV

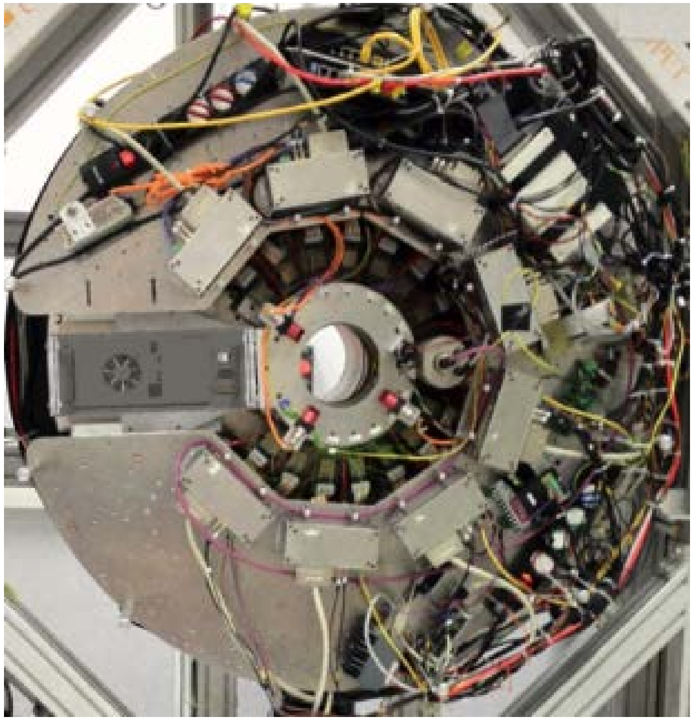
ClearPET



Images acquired with ClearPET in Julich, Germany

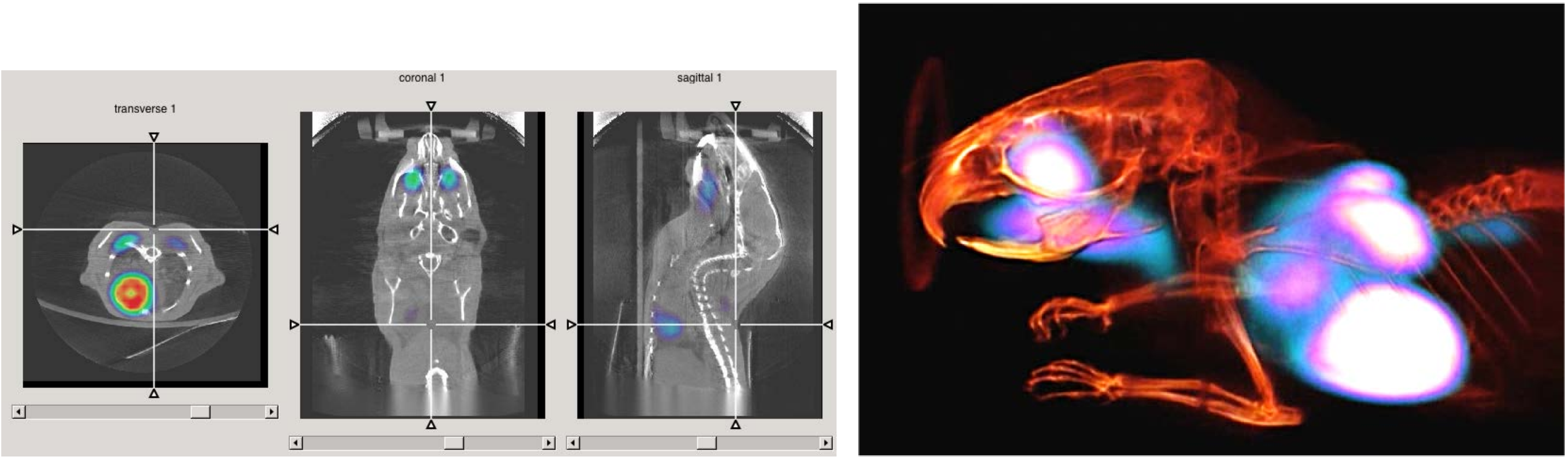
ClearPET/Xpad @ CCPM





Simultaneous PET/CT

ClearPET/Xpad @ CCPM



Simultaneous PET/CT

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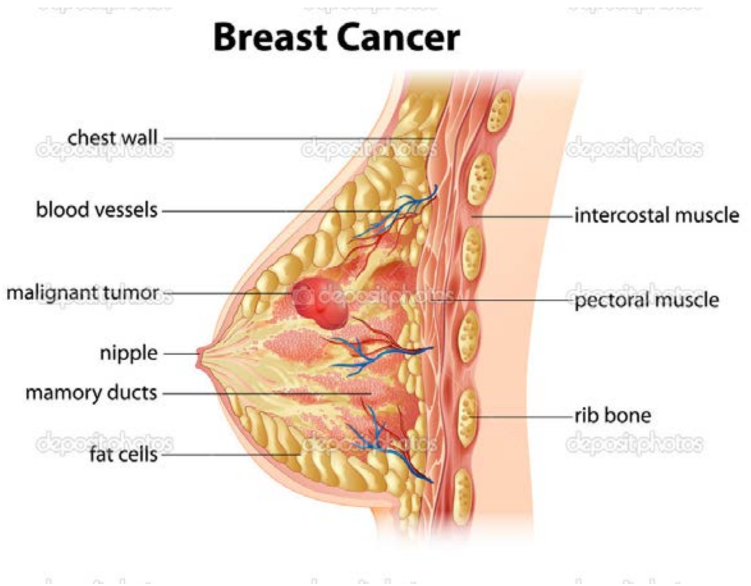
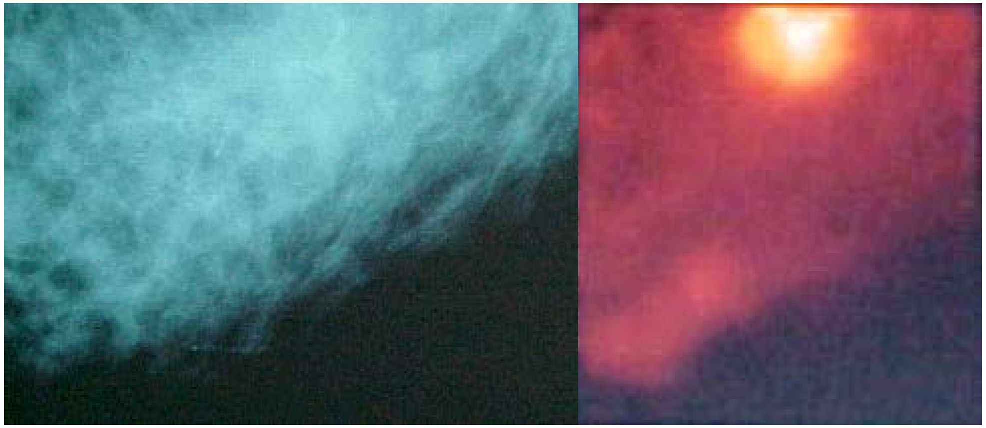
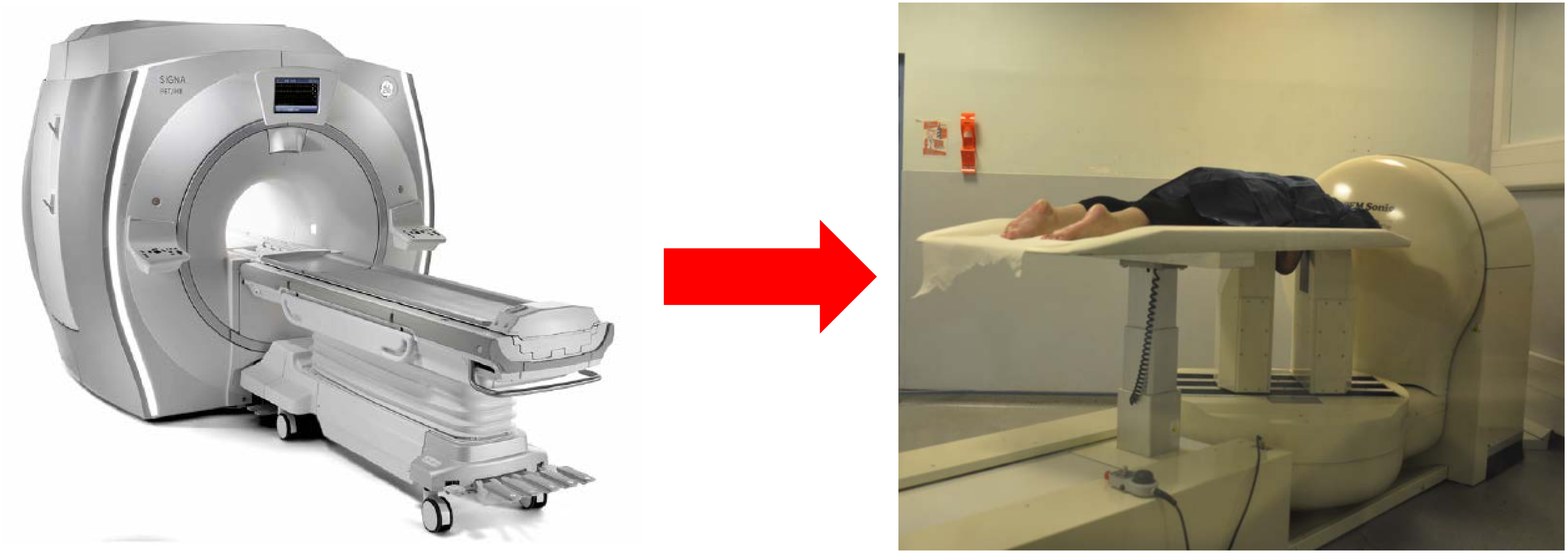
ClearPEM: motivation





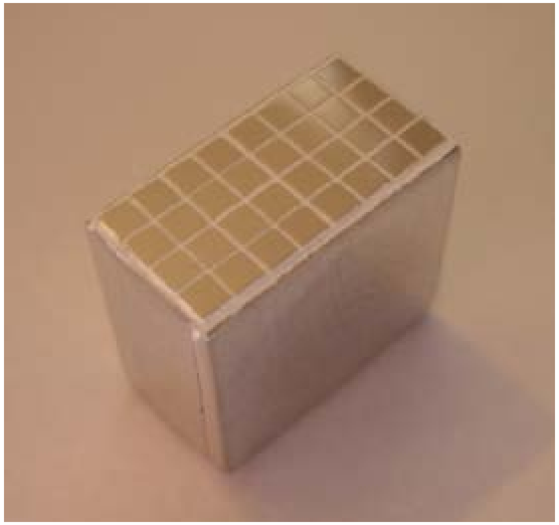
- **Target**: breast cancer
  - 1 over 8 women affected during lifetime, 2nd cancer-related cause of death in women
  - Survival rate dramatically improves if diagnosed in early stage
- **Issues**: standard detection techniques (X-rays) suffer from low specificity
- **Goal**: improve sensitivity and spatial resolution (for breast)
  - Better specific **sensitivity** for breast
  - Very high **spatial resolution**

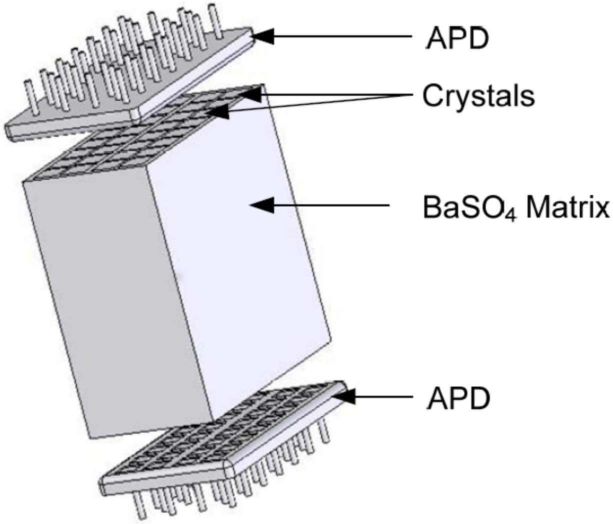
ClearPEM: sensitivity



From whole body to organ dedicated scanner

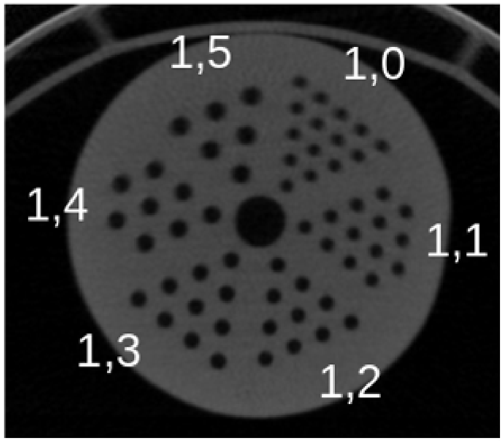
ClearPEM: spatial resolution

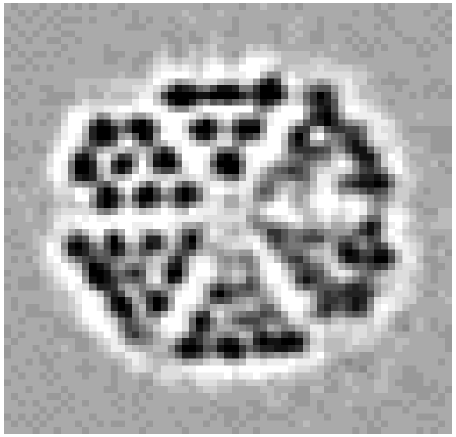




- Small section of individual scintillators ($2x2x20mm^3$)
  - Spatial resolution < 2 mm
- Measurement of DOI
  - Implemented with double side readout of each scintillator
  - Correction of parallax errors
  - Improved spatial resolution and homogeneity

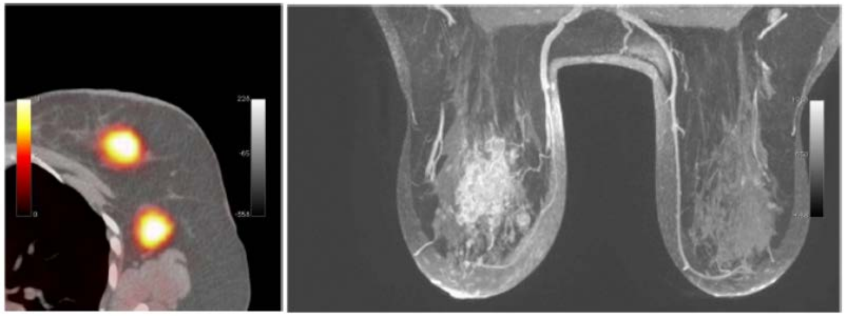
ClearPEM: performance





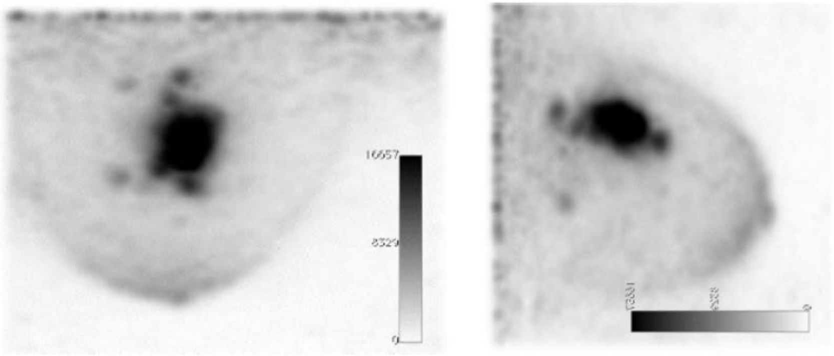
Parameter	Value
Spatial resolution FWHM	~1.4 mm
Energy Resolution FWHM	15.5%
CTR res. FWHM	2.8 ns
DOI res. FWHM	3 mm

Patient sample image - 1



Whole-body PET/CT

MRI

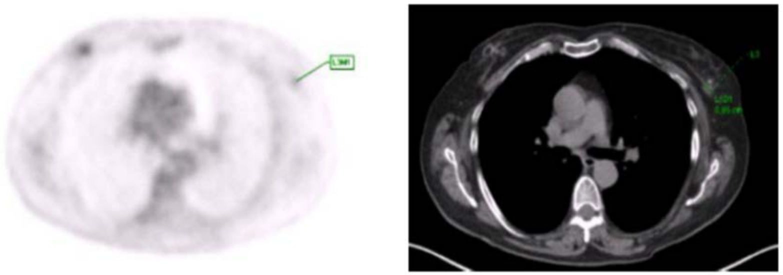


Coronal ClearPEM-Sonic

Sagittal ClearPEM-Sonic

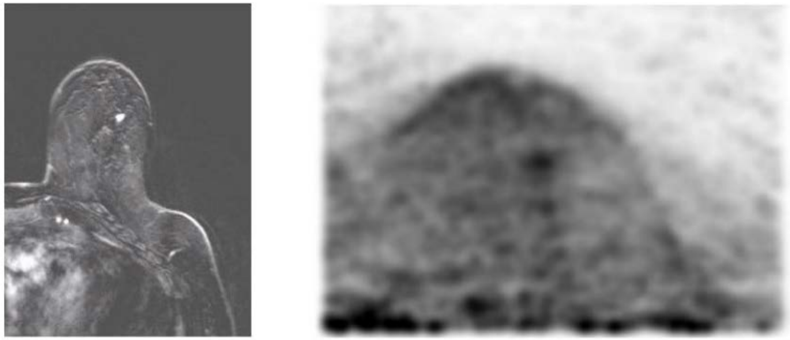
Ability to resolve multifocal lesions

Patient sample image - 2



Whole-body PET

CT



MRI

ClearPEM-Sonic PEM

Detection of lesions invisible to WB-PET and CT

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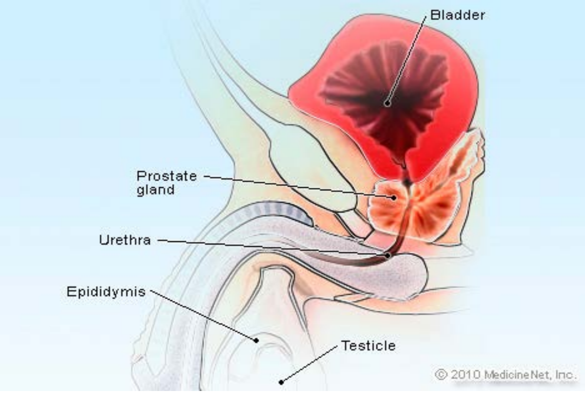
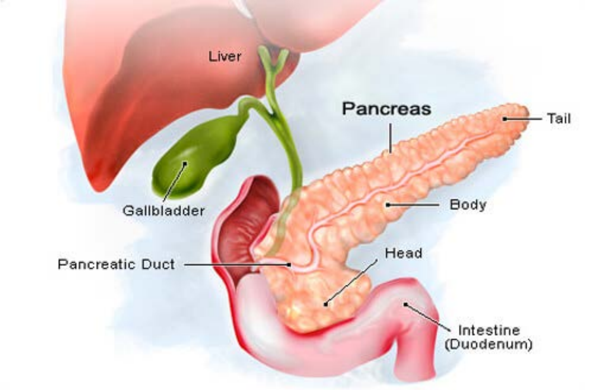
EndoTOFPET-US

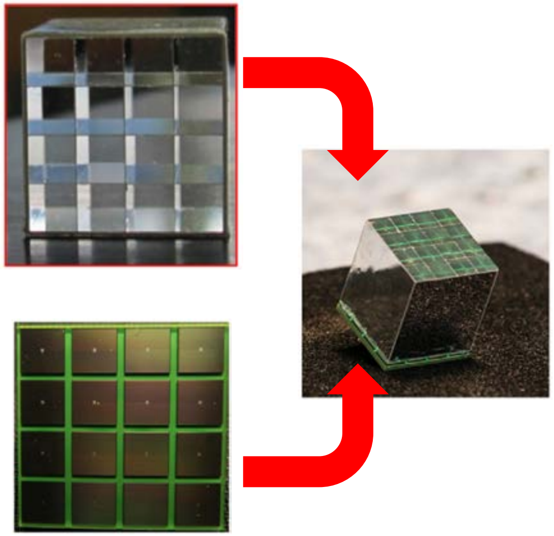


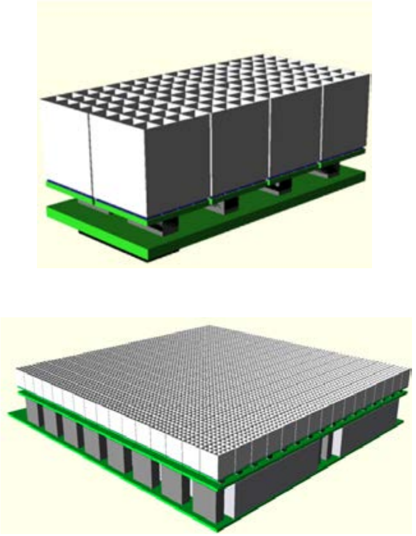
© 2010 MedicineNet, Inc.

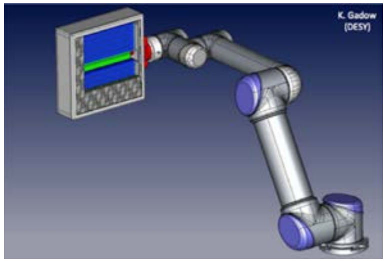


- **Target**: prostate and pancreas cancer
  - Prostate cancer: very **common** in men
  - Pancreatic cancer: very **aggressive**, difficult diagnosis
- **Issues**: currently PET/CT do not provide early diagnosis
  - Small lesions, below detectability until advanced stage
  - Strong background from neighbouring organs (heart, liver, bladder)
- **Goal**: achieve high spatial resolution and reject background
  - **Endoscopic** approach
  - Time of flight (TOF) around **200 ps -> 3cm**

EndoTOFPET-US: external plate

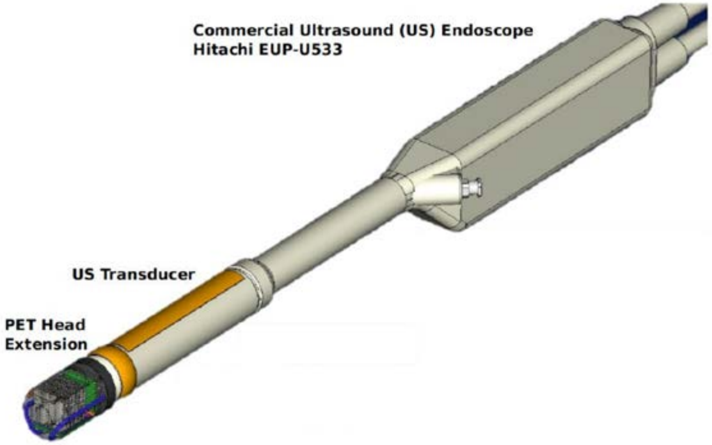


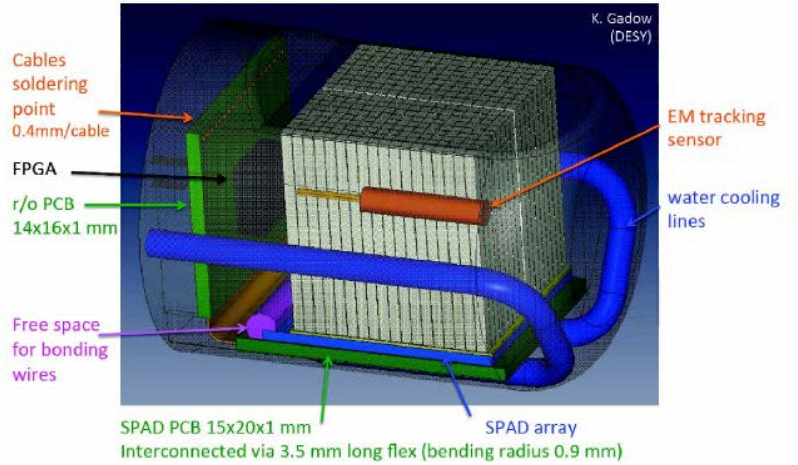




- **Two plates** produced (one for prostate detector, one for pancreas detector)
- 256 arrays of 4x4 **LYSO:Ce** scintillators for each plate
  - ➢ Individual crystal size: **3.5x3.5x15 mm^2** for prostate, **3.1x3.1x15 mm^2** for pancreas
  - ➢ Crystal pitch: **3.6 mm** for prostate, **3.2 mm** for pancreas
- Discrete Silicon-through-via (TSV) **MPPCs** by Hamamatsu, RTV 3145 glue
- **FEB/A** with 8 modules and 2x64ch readout **ASICs**, 4 **FEB/D** with 8 FEB/A each
- Cooling system, mechanical arm

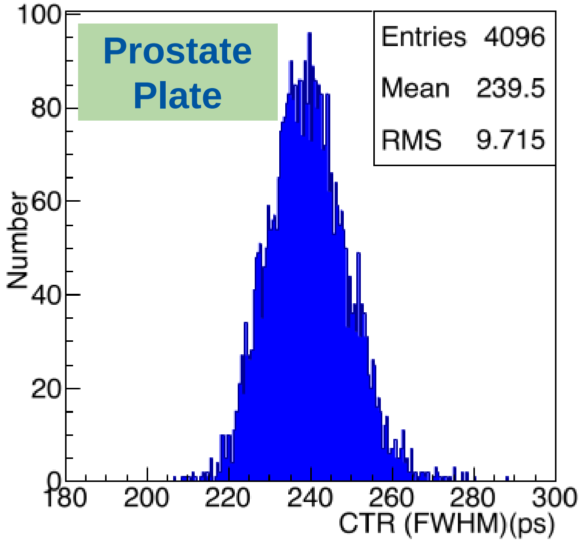
EndoTOFPET-US: probe

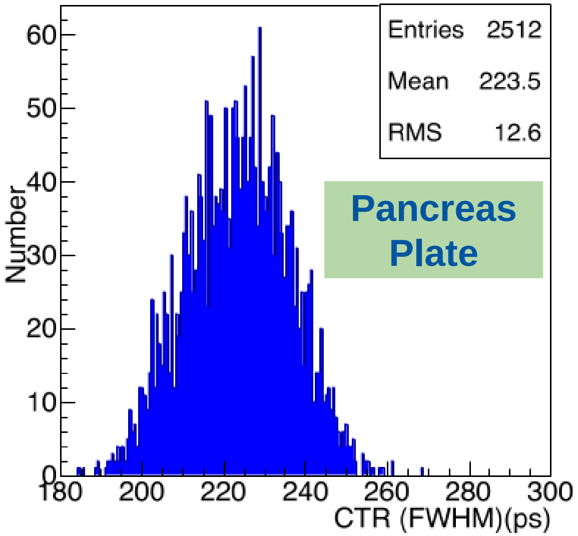




- **Two different versions** under development:
  - Pancreas probe, diameter **15 mm**
  - Prostate probe, diameter **23 mm**
- **Scintillators**: 1 (pancreas) or 2 (prostate) arrays of **9x18 LYSO:Ce**
  - Individual crystal size **0.71x0.71x15** (or 10) **mm^3**
- **Photo-detector**: custom **MD-SiPM** developed within the collaboration
- EM, and optical **tracking**, water **cooling**

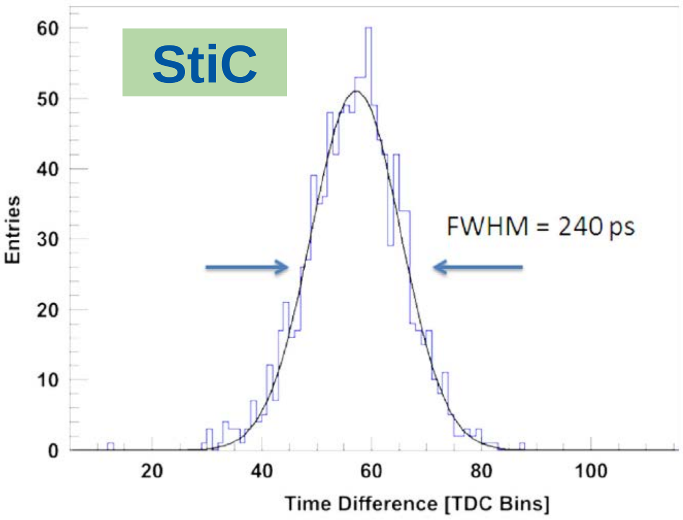
Performance: modules

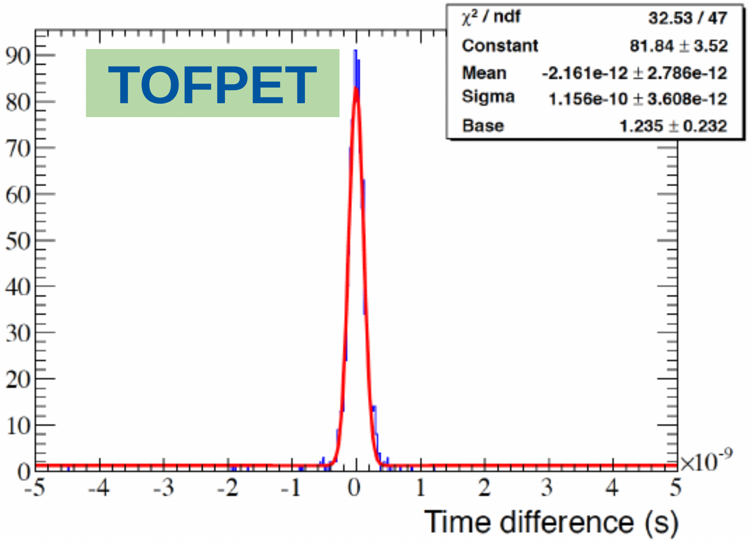




- Light Output of all modules determined as number of pixels fired
  - ➢ Module excited with ^{22}Na source
  - ➢ Current output integrated by QDC over 100 ns gate
  - ➢ Mean Light Output = 1876 +/- 100 pixels fired
  - ➢ Mean **Energy Resolution FWHM** = **12.8%**
- Coincidence Time Resolution (CTR)
  - ➢ Measured with NINO and HPTDC for each module against a reference module
  - ➢ Average prostate plate **CTR$_{\mathbf{FWHM}}$** = **239.5 ps**
  - ➢ Average pancreas plate **CTR$_{\mathbf{FWHM}}$** = **223.5 ps**

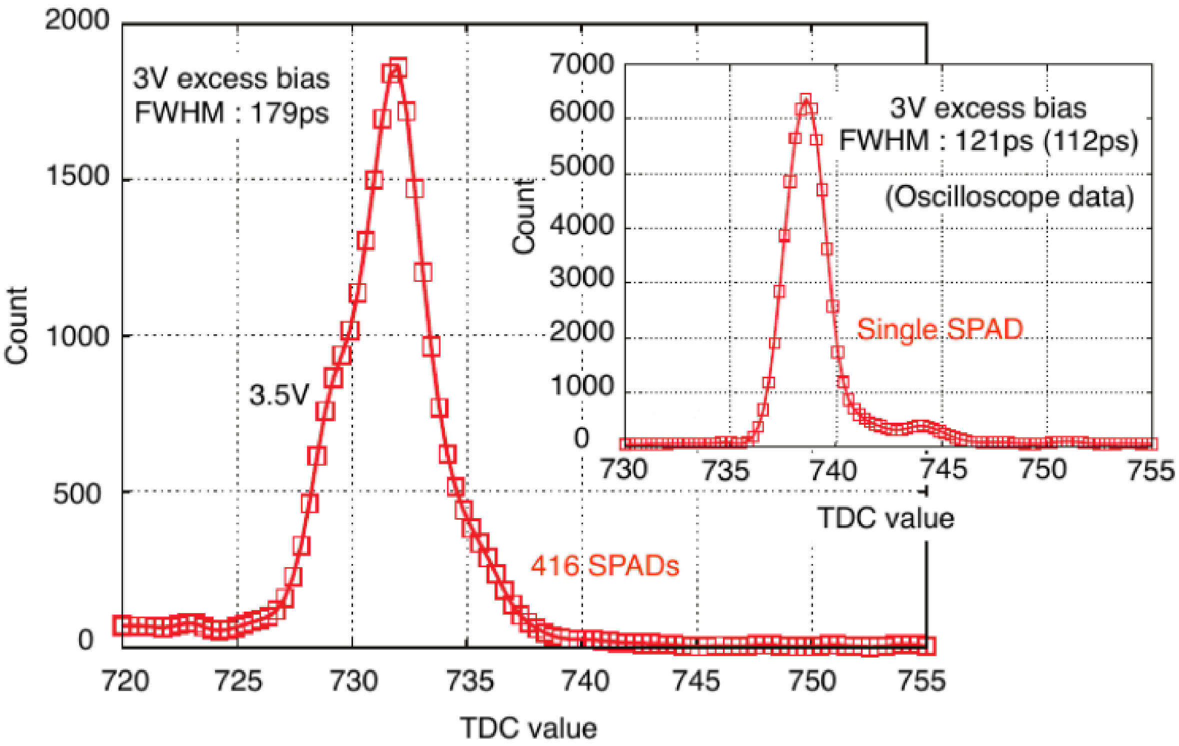
Performance: ASICs





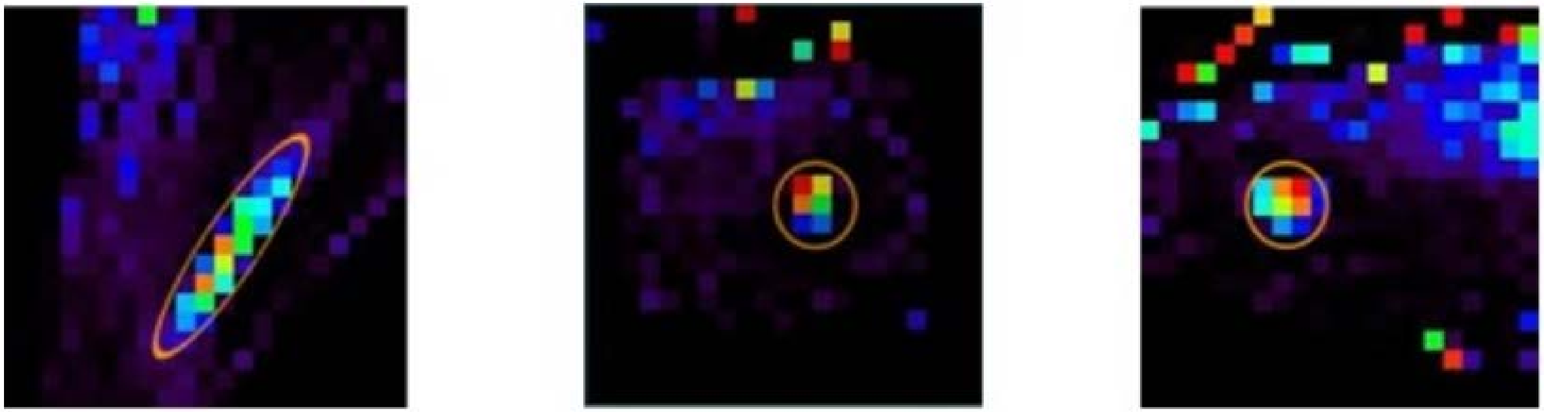
- Two dedicated fast 64 channel ASICs developed: **StiC** and **TOFPET**
  - Leading edge technique to get timing information
  - Linearized Time-Over-Threshold method to provide energy information
  - Low noise, low timing-jitter, low power consumption
- **CTR** measured for both ASICs
  - Single 3.1x3.1x15 mm^3 crystals coupled to 2 Hamamatsu MPPCs
  - ^{22}Na source
  - StiC average **CTR**$_{\mathbf{FWHM}}$ = **240 ps**
  - TOFPET average **CTR**$_{\mathbf{FWHM}}$ = **270 ps**

Performance: MD-SiPM



- Single Photon Timing Resolution (SPTR) evaluated
- **$\mathbf{SPTR_{FWHM}}$** evaluated in **121 ps** for single SPADs and **179 ps** for entire detector

Reconstruction algorithm



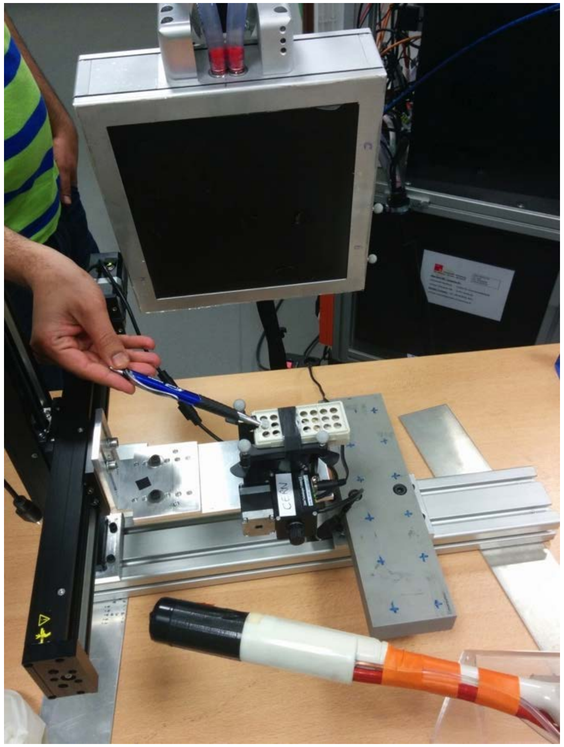
Transverse

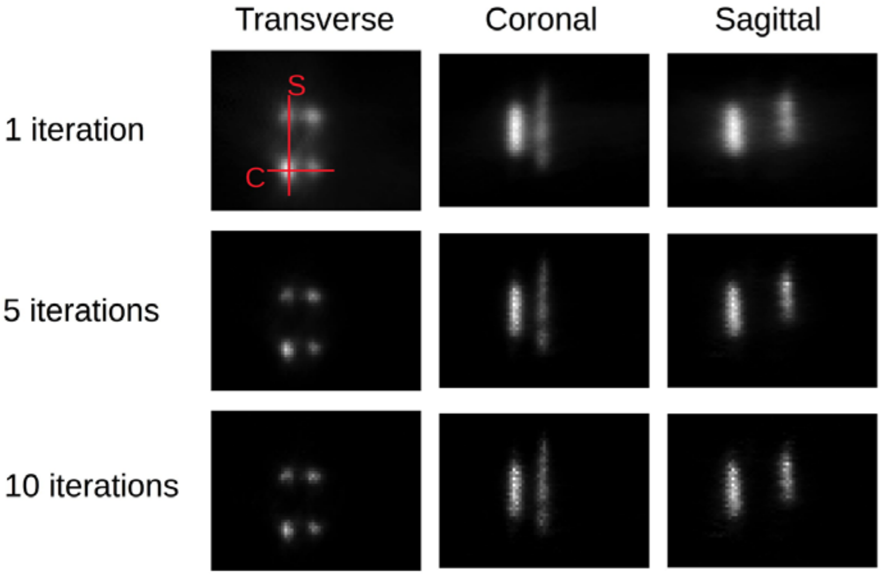
Coronal

Sagittal

- Dedicated reconstruction algorithm developed within the collaboration
  - Iterative **histogram based ML-EM** reconstruction
  - Incorporates TOF information
  - Copes with detector asymmetry
  - Takes into account the limited rotation capabilities
- Expected performance tested on **simulated datasets**
  - Based on GAMOS Monte Carlo toolkit
  - **1 mm resolution** within reach with **10 minutes** scan time

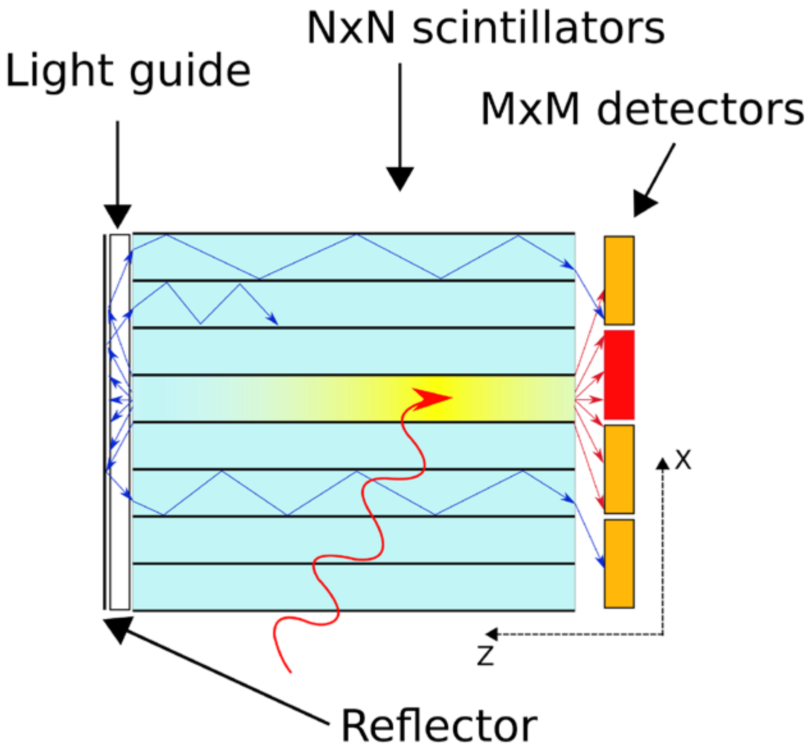
Testing of first prototype





- **Provisional probe** with 2 MPPCs and 2 4x4 LYSO:Ce arrays (3.1x3.1x15 mm^3)
- Clamping on prostate US endoscope
- Preliminary images obtained at **CERIMED-Marseille** on cylinders filled with **FDG**

Example of current research: DOI and TOF



- DOI information and multiple timestamps to **correct transit time jitter**
- See presentation **O15-Tue** from A. Polesel on Sept. 11th

Conclusions

- PET is a fundamental **molecular imaging technique**

- Detectors resolution **limits**

➢ Spatial resolution ->	geometrical, mathematical, intrinsic
➢ Energy resolution ->	scintillators
➢ Timing resolution ->	scintillators, detectors, electronics

- Various PET developed within **Crystal Clear Collaboration**
  - ➢ **ClearPET**: pre-clinical scanner for animals
  - ➢ **ClearPEM-Sonic**: clinical mammography PET
  - ➢ **EndoTOFPET-US**: endoscopic PET for prostate and pancreas

Acknowledgements

CRYSTAL CLEAR

CERN

ENDO TOFPET US
Endoscopic TOFPET & Ultrasound

FAST
Fast Advanced Scintillator Timing

TICAL

THE PICOSEC
MC-NET PROJECT

KT-Fund
CERN

ASCIMAT

Radiotracer

- Principle of **radiotracer** (1943):

> *“the changing of an atom in a molecule with its radioisotope will not change its chemical and biological behavior”*

Radiotracer

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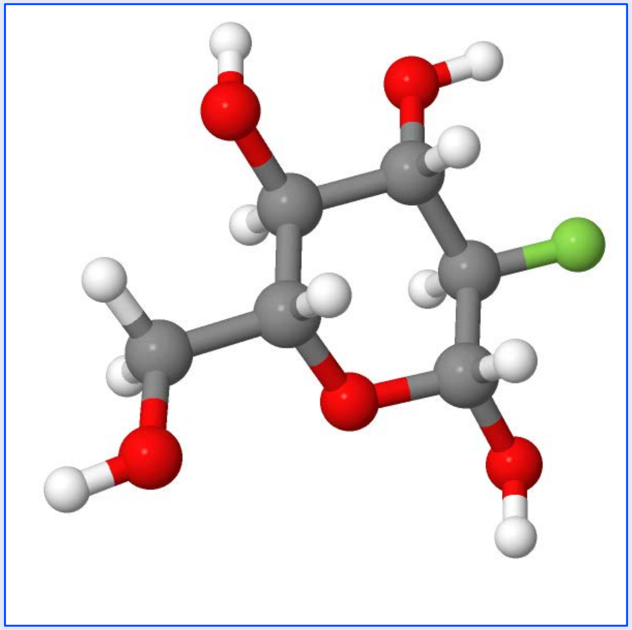
- An example: **cancer** and the **Warburg effect (1924)**
  - While healthy cells produce energy by moderate glycolysis followed by oxidation of pyruvate mitochondria…
  - … in cancer cells the glycolysis rate is much higher, followed by lactic acid fermentation
  - As a result, the **uptake of glucose by cancer cells** is much higher

- With the proper radiotracer, it is possible to gain information on specific **metabolic processes**

Common radiotracers

- **Oncology**
  - ^{18}F-FDG (glucose metabolism)
  - ^{11}C-Thymidine (DNA synthesis)
  - ^{18}F-FLT (DNA synthesis)
- **Neurology**
  - ^{18}F-FDG (glucose metabolism)
  - ^{18}F-DOPA (dopamine)
- **Cardiology**
  - Rubidium-82 (potassium analog)
  - ^{13}N-Ammonia (myocardial perfusion)
  - 15-O_2 (oxygen cycle)



Fludeoxyglucose ***(^{18}F-FDG)***